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When exposed to challenged ventilation, those with a history of LBP increase spine stability relatively more than healthy individuals

S.G. Grenier [a,*], S.M. McGill [b]

[a] *Faculty of Professional Schools, School of Human Kinetics, Laurentian University, Sudbury, Ontario, Canada P3C 1N8*
[b] *Faculty of Applied Health Sciences, Department of Kinesiology, University of Waterloo, Waterloo, Ontario, Canada*

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Abstract

Objective. To determine if spine stability would be affected by the competing demands of simultaneous challenged ventilation and supporting a hand-held load.

Design. Subjects were their own controls in a repeated measures design where a single task was repeated, once in a different condition, in a random order.

Background. Muscle stiffness influences spine stability. The same muscles that contribute to spine stability assist in challenged breathing. We hypothesized that a challenged ventilation task would place low back pain (LBP) sufferers at risk of spine instability.

Methods. Subjects (14 normal; 14 with low back pain) performed two trials with a 22 kg hand-held weight and the trunk angled forward at 30°. One trial was of 60 s duration while breathing ambient air, the other of 70 s duration, while breathing 10% carbon dioxide. Spine stability and compression were quantified, using an EMG assisted optimization model in both trials.

Findings. Contrary to expectation, spine stability increased during the challenged breathing trials compared to the ambient air condition for subjects with a history of low back pain when abdominal muscle activity was accounted for as a covariate.

Interpretation. Subjects with a history of low back pain had higher stability in challenged breathing trials, indicating that some active mechanism protects the spine for the LBP groups in challenging situations. This may be to provide some margin of safety for damaged passive tissues but could be adversely affected by fatigue in the longer term.

Keywords: Spine stability; Low back pain; Motor control; Spine load; Ventilation

1. Introduction

Spine stability, during isometric holds is dependent upon symmetric muscle activity levels, balanced against an external load (Cholewicki and McGill, 1996; Stokes et al., 2000). Co-contraction in general (Granata and Marras, 2000), and of the abdominals specifically (Gardner-Morse and Stokes, 2001; Granata and Marras, 2000; Kavcic et al., 2004a) has been shown to increase torso stiffness and stability in the lumbar spine. McGill et al. (1995) have documented that, during quiet breathing (even when supporting a heavy load), it is normal to have very little entrainment of abdominal wall muscle to ventilation, since a healthy diaphragm and lung elasticity seem sufficient for proper ventilation. However, during challenged breathing, when the diaphragm contraction draws air for inspiration, the abdominal muscles are often recruited to assist with elastic recoil by "active expiration" (Abraham et al., 2002; Aliverti et al., 2002). This process represents a form of co-contraction which should affect spine stability. The rationale for this study is the paradox presented to the motor control system: that the abdominal muscles recruited for breathing are also required to maintain spine stability with an isometric contraction (McGill et al., 1995). Will spine stability suffer if the motor control system must meet the simultaneous demand of challenged breathing and maintaining a load? Given the muscle co-activation

* Corresponding author.
E-mail address: sgrenier@laurentian.ca (S.G. Grenier).

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requirements that spine stability places on the motor control system (Granata and Wilson, 2001) it seems unlikely that both ventilation and stability could be optimally achieved. Interestingly, similar situations exist in other animals. For example, Owerkowicz et al. (1999) report that, in monitor lizards, mechanical ventilation requirements are a limiting factor to locomotion velocity. This study investigates the possibility that a similar coupling exists between ventilation and the need for a stable spine in humans. This would have implications for both injury risk and performance potential.

It has long been established that those with a history of injury have a greater risk of subsequent injury (McGill, 2007). In recent years, low back injury has become more firmly associated with neuromuscular deficits, possibly affecting the ability to properly stabilize the spine summarized in McGill (2007), and in Richardson et al. (2004)). While the causative relationship remains unclear, recent work suggests that, in the motor control system, plasticity exists for adjustments in muscle co-contraction to maintain stability (Granata and Wilson, 2001). Interestingly, this co-activation, required for stability, also increases spine compression.

However, the benefit of increased stability far outweighs the increased cost of compression by approximately 3–1 (Granata and Marras, 2000) and compression, itself, helps to stabilize the spine (Janevic et al., 1991). Further, Cholewicki and McGill (1996), demonstrated that stiffening the muscles to increase elastic potential energy directly increases stability. More recently, Kavcic et al. (2004a) have shown that simultaneous activation of moment antagonists also enhances stability. The co-contraction response to postural and purely mechanical changes has been demonstrated in a simplified model where stability was a requirement (Granata and Wilson, 2001). The stabilizing response of a more complex neuromuscular system to more complex tasks with conflicting demands is, to this point, poorly understood.

The purpose of this study was twofold (1) to better understand the effect of conflicting demands on spine stability and (2) to establish if there is a different response in those with low back pain. More specifically, a situation was explored where the muscles contract and relax to assist with challenged breathing yet must also contract to support a hand-held load. It was hypothesized that those subjects with a prior history of LBP would sacrifice spine stability to maintain ventilation in the face of competing demands.

2. Methods

2.1. Data collection

This study was approved by the University Human Research Ethics Committee and all subjects provided informed consent. Workers from physically demanding jobs ($n = 28$) volunteered for this study. While additional people volunteered, those with current low back pain were excluded from participating. Those subjects who had low back pain (whether they missed work or not) within the last year (yLBP, $n = 14$) were categorized apart from those with no history (nLBP, $n = 14$). Subjects with no low back pain (nLBP) were on average 37.5 years of age (SD 8.12), 1.76 m tall (0.079), and had a mass of 80.6 kg (10.9). Subjects with a history of low back pain (yLBP): 36.4 years (8.14), 1.82 m (0.065) and 92.4 kg (12.0). Lumbar spine kinematics were recorded with a 3 Space Isotrak unit (Polhemus, Colchester, VT, USA) which sampled three axes of instantaneous spine motion (flexion–extension, lateral bend and axial twist) at a rate of 60 Hz. The electromagnetic field source of the Isotrak was strapped over the sacrum and a sensor was worn over the twelfth thoracic vertebrae. Electromyographic (EMG) signals were recorded using bipolar surface electrodes 25 mm apart at 1024 Hz from seven channels bilaterally (14 total): rectus abdominis (RA: 2 cm lateral to the umbilicus), internal oblique (IO: below the anterior superior iliac spine (ASIS) but above the inguinal ligament, external oblique (EO: approximately 15 cm lateral to the umbilicus positioned obliquely in line with the fibres), latissimus dorsi (15 cm lateral to T9 positioned obliquely in line with the fibres), thoracic erector spinae (TES: 5 cm lateral to T9 over the muscle belly), lumbar erector spinae (LES: 3 cm lateral to L3) and the multifidus (MF: 2 cm lateral to L5, angled slightly with the superior electrode more medial). The collected signals were A/D converted at a sample rate of 1024 Hz (frequency response: 10–1000 Hz, common mode rejection ratio: 115 dB at 60 Hz, input impedance: $\sim$10 GΩ) and normalised to the amplitudes measured during the maximum voluntary contraction (MVC) procedure following rectification and low pass filtering at 2.5 Hz. The MVC procedure involved the subjects performing maximum isometric (resisted by experimenter) effort exertions in flexion, extension and twisting tasks in an attempt to elicit maximum electrical activity (described in detail in McGill, 1991). An ultrasonic flow meter (model #UF202, Kou Engineering, NOVEX, Redmond, WA, USA), in line with the mouthpiece, also sampling at 1024 Hz, recorded ventilation flow rate.

Subjects performed two isometric weight holding trials (22 kg) of 60 s duration; one while breathing ambient air and the other while breathing 10% carbon dioxide. The subject's arms hung perpendicular to the floor and they lifted the weight until the amount of trunk flexion was at approximately 30°, controlled by a physical guide. Knee flexion was not controlled. Subjects were told to adopt a comfortable posture without locking their knees but keeping their feet stationary and shoulder width apart. On average the EMG assisted model (Cholewicki and McGill, 1996) calculated an average moment of 112.5 Nm, and a compressive load on the L4/L5 joint of about 2400 N, well below the NIOSH action limit (Waters et al., 1993). The main reasons for choosing this were, to have more control over exertion, to still stress ventilation and the associated muscles while keeping the task relatively static so that any difference in stability could more easily be associated with specific contributors to stability.

2.2. Stability model

The model used in this experiment has been fully described elsewhere (Cholewicki and McGill, 1996). A brief description of some recent improvements is provided here. Improvements were made to better represent transversus abdominis with two vertebral attachments pulling laterally via the superficial (tip of the posterior spinous process) and deep fascia (transverse process). Four fascicles of quadratus lumborum were added which originated on the transverse processes of L5–L2 and attached to the ribs (Bogduk et al., 1992). The cross-sectional areas of multifidus and pars lumborum were adjusted so that the physiological area at each level closely approximated previous findings from MRI scans (McGill et al., 1993).

During torso bending, stiffness resulting from stressing passive tissues creates a passive moment which counteracts the external moment. In both flexion and lateral bend the moment created by this stiffness was adjusted based on the range of motion of each individual's spine (Cholewicki and McGill, 1996). A pre-load bias component was added which accounted for the increase in torsional stiffness for each increase in compression. An exponential function was fitted to data from osteo-ligamentous spines given by Edwards et al. (1987) and Janevic et al. (1991), where the passive moment was adjusted by the interaction of angle and preload.

$$
\begin{aligned}
M_x &= A \cdot e^{k\theta} + B(\mathrm{CMP}) \\
M_y &= A \cdot e^{k\phi} + B(\mathrm{CMP}) \qquad (1) \\
M_z &= A \cdot e^{k\psi} + B(\mathrm{CMP})
\end{aligned}
$$

See Table 1 for value of coefficients. No negative angles are input to these equations. For example right bend or left bend are both considered positive. θ, ϕ, ψ are angles in radians about the respective axis of rotation.

The passive stiffness was combined with active stiffness generated by the muscles and calculated using the moment distribution method (Cholewicki and McGill, 1995). Stability was then quantified as is described in Cholewicki and McGill (1996). The stability index, where zero or negative values are unstable and smaller positive values are less stable, was given by the determinant of the diagonalized Hessian matrix of the second partial derivatives of potential energy relative to the Euler angles which specify the rotation angles of the spine (Cholewicki and McGill, 1996). While other criteria have been used in the past (eg, minimum eigenvalue), the comparative result changes little (Howarth et al., 2004). The mean of 18 eigenvalues is highly correlated with the magnitude of the smallest eigenvalue.

Table 1
The coefficients for Eq. (1) for each of the three orthogonal planes of motion

Direction	A	k	B
Flexion (M_{z-})	1.2069	1.287	0.0018
Extension (M_{z+})	5.213	0.6103	0.0006
Lateral bend (M_x)	1.2074	1.288	0.0024
Axial twist (M_y)	3.3404	24.53	0.0016

Note: in each case CMP is the spine compression value.

2.3. Data analysis

Stability was evaluated using the mean stability index over the duration of the trial. A within subject mixed design ANOVA with one repeated measure (CO_2, ambient air) and one between variable (yLBP, nLBP) was used to distinguish whether stability measures differed, either across groups or conditions, or whether an interaction existed between LBP group and breathing condition. In all cases the independent variables were the breathing conditions (ambient air and CO_2) and the groups were split based on LBP history. A repeated measures, within subject ANOVA, testing if each muscle's EMG root mean square (RMS) average differed, either across ventilation type or LBP group, was also performed for both abdominal and erector spinae muscles. In this analysis each muscle's coefficient of variation was calculated and used as a covariate. An ANOVA was also done on the cross-correlations between each muscle and the stability output from the trial of interest. The dependent variable was the difference between the maximum cross-correlation output and the minimum cross-correlation output, to account for the degree of synchronization between EMG and stability. The factors were LBP group and ventilation type.

3. Results

Subjects with a history of LBP did not sacrifice stability in order to meet ventilation demands; their stability actually increased. It was expected that the yLBP group would have lower stability values when exposed to a ventilatory challenge, but the opposite effect was observed.

While there were other differences between groups in our results, these tended to be between ambient air and CO_2 trials rather than between LBP groups. The RMS EMG (Table 2) of both the right and left rectus abdominis and external obliques were different between ambient air trials and CO_2 trials but not between LBP groups. There were no differences in lumbar compression values between either LBP groups or trial types (ambient or CO_2) (Fig. 1).

Upon further exploration, trial type (ambient air or CO_2) and LBP category (nLBP or yLBP) interacted to significantly affect stability index when right ($P = 0.04$) and left ($P = 0.03$) rectus abdominis were used as covariates in the ANOVA. For those with a history of LBP, the stability index actually increased ($P < 0.05$) when they were exposed to CO_2 whereas in those who had never had LBP the stability index decreased when they were exposed to CO_2 (Table 3, Fig. 2).

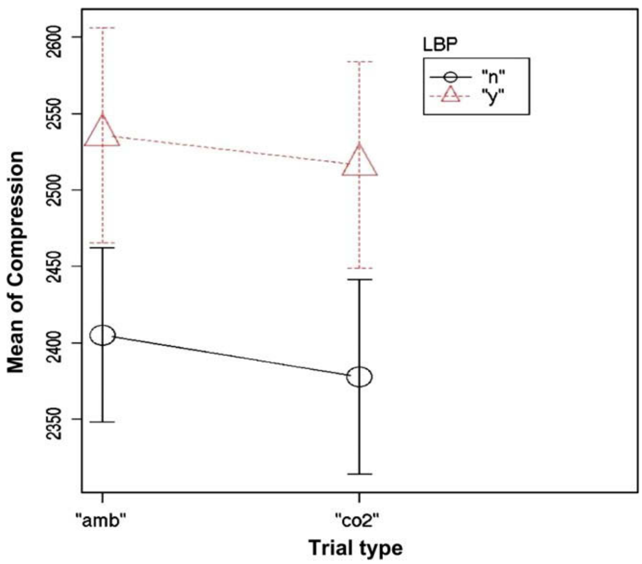

Fig. 1. This plot shows that, while not significant the yLBP group had greater lumbar compression under both experimental conditions. In both groups compression decreased when exposed to CO_2. This is important because, in spite of compression not being significant, the cost benefit ratio (stability to compression) goes in opposite directions for the two groups when exposed to CO_2. The ratio goes up for yLBP when exposed to ambient compared to CO_2) (0.46 vs. 0.47) while nLPB ambient compared to CO_2 reverses (0.47–0.46).

To verify that the experimental condition had the same effect on both groups, an ANOVA was performed with ventilation rate as the dependent variable and LBP group and ventilation type as the factors. There was no difference in ventilation rate between LBP groups but there was a difference between ventilation types ($P = 0.0001$). There were no differences in mean trunk flexion angle between conditions or groups.

An ANOVA was performed on the cross-correlations differences between individual muscles normalized and filtered EMG and the stability profiles as the dependent variable and LBP group and ventilation type as the factors. While the correlations were generally very low (ranging from −0.47/LBP/CO_2 to 0.27/nLBP/AMB), the result was a significant difference in correlation between both LBP groups ($F = 10.59$, $P < 0.05$) and ventilation groups ($F = 11.22$, $P < 0.05$). The interaction between LBP group and vent group was not significant ($F = 3.36$, $P = 0.06$). Those in the LBP group had a lower average correlation (more negative) in the CO_2 trials (four times more than LBP) while the nLBP group had higher average correlation (more positive) in the AMB trials (two times higher). A chi squared analysis, counting individual cases of significant correlation between ventilation pattern and stability index, was significant (Table 4). With CO_2 exposure, the yLBP group tended to go from ventilation being uncorrelated to stability to a significant correlation pattern between stability and ventilation.

Since the optimization process modified the EMG to balance the internal moments against the external moments

Table 2
The mean RMS EMG values (as a percentage of MVC) for each trial for all muscles that were input to the stability model

	LBP	AMB		CO_2	
		Mean	SD	Mean	SD
Left_RA	n	2.17	1	2.28	1.05
	y	2.75	2.64	2.92	2.66
Left_IO	n	2.33	1.61	2.31	1.48
	y	3.13	2.16	3.19	2.21
Left_EO	n	2.69	3.26	2.87	3.29
	y	3.41	2.5	3.77	2.97
Left_LT	n	14.59	10.11	13.58	8.61
	y	8.5	6.21	6.92	6.03
Left_MF	n	17.74	7.17	17.31	7.9
	y	20.41	7.73	20.24	8.27
Left_LE	n	15.44	7.54	15.53	8.28
	y	15.21	5.58	14.86	5.56
Left_TE	n	15.67	4.89	15.73	4.27
	y	16.21	5.91	15.01	5.59
Right_RA	n	1.75	0.67	1.86	0.81
	y	2.23	1.51	2.35	1.53
Right_IO	n	2.57	2.12	2.27	1.77
	y	3.19	2.11	3.08	2.34
Right_EO	n	2.07	2.02	2.25	2.02
	y	2.24	1.76	2.55	2.02
Right_LT	n	14.83	11.49	11.87	6.69
	y	9.01	6.55	7.85	7.29
Right_MF	n	18.55	6.78	18.51	7.96
	y	17.8	7.42	17.46	7.23
Right_LE	n	12.89	3.82	12.89	5.09
	y	15.44	3.44	15.83	4.59
Right_TE	n	16.29	4.99	16.99	4.09

The means are categorized, for each muscle, by the type of trial (AMB = ambient air, CO_2 = carbon dioxide) and whether it was from a person with low back pain (yes or no).

the mean changes to those moments were monitored. The mean gain applied to the muscles, split by groups was for

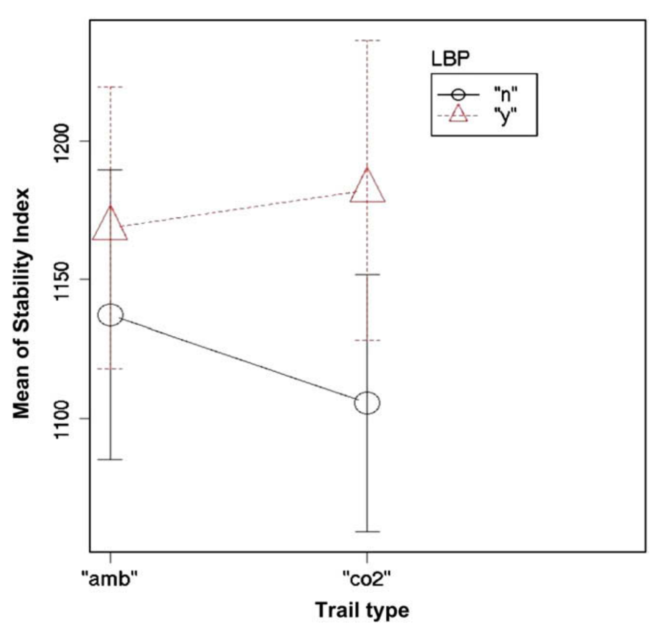

Fig. 2. Although this plot does not account for the covariates of muscle activity, the interaction between breathing condition and LBP group when measuring spine stability is clear. An ANCOVA confirmed this effect.

Table 3
Mean values for the measures (dependent variables) of integrated stability index, mean stability index, as well as lumbar flexion angle and lumbar compression

Measure (DV)	Low back	Ventilation	Mean	St dev
Stability index (N m/rad^2)	yLBP	Ambient	1168.7	183.3
Stability index	yLBP	Challenged	1182.5	195.0
Stability index	nLBP	Ambient	1137.4	189.0
Stability index	nLBP	Challenged	1105.4	167.6
Flexion angle (deg)	yLBP	Ambient	30.8	10.9
Flexion angle	yLBP	Challenged	30.7	13.3
Flexion angle	nLBP	Ambient	32.5	11.5
Flexion angle	nLBP	Challenged	35.1	11.5
Compression (N)	yLBP	Ambient	2535.9	253.5
Compression	yLBP	Challenged	2516.5	243.3
Compression	nLBP	Ambient	2405.2	205.8
Compression	nLBP	Challenged	2377.6	229.4

Values are given for ventilation conditions of ambient air, challenged breathing as well as for cases of yLBP and nLBP.

Table 4
For each combination of conditions the number of significant correlations, between flexion angle and stability index, were tabulated

	LBP ambient	LBP CO_2	nLBP Ambient	nLBP CO_2
No correlation	7	2	7	5
Negative correlation	3	4	4	4
Positive correlation	3	7	3	5

A chi square analysis of these frequencies was not significant regardless of pooling, such that effect of flexion angle on stability had no consistent effect.

Table 5
For each combination of conditions the number of statistically significant correlations, between ventilation pattern (from the flow meter) and stability index, were tabulated

	LBP ambient	LBP CO_2	nLBP Ambient	nLBP CO_2
No correlation	10	5	12	9
Negative correlation	2	5	0	1
Positive correlation	1	3	2	2

A chi square analysis of these frequencies was significant when positive and negative correlations were pooled, such that those with no low back pain were more often able to dissociate ventilation from stability.

lbp − amb = 1.24 (SD = 0.44); lbp-CO_2 = 1.19 (SD = 0.41); nlbp − amb = 1.18 (SD = 0.39); nlbp − CO_2 1.15 (SD = 0.56). There were no significant differences ($P = 0.7$) among any of the groups in the gains applied to the EMG (see Table 5).

4. Discussion

Contrary to our hypothesis, nLBP subjects showed a relative decrease in stability levels when exposed to CO_2 while yLBP subjects showed an increase in stability (Fig. 2. Four other results help to shed light on this main finding. First, the cross-correlation of EMG to stability are different between LBP groups. The nLBP were higher during the ambient air condition and lower in the CO_2 conditions, but the opposite was true for yLBP. This follows the same pattern as the main result. Second, the yLBP group more often showed a pattern of correlation (either positive or negative) between stability and ventilation. Third, mean trunk flexion angle was not significantly different, either between LBP groups or ventilation conditions. Finally, there was a significant interaction between ventilation condition and LBP group when RA was applied as a covariate. These results mean that the difference in stability cannot be accounted by a difference in trunk flexion angle, increasing stability with a greater angle, because there was no difference in angle between either groups or conditions. Nor can it be accounted for by a difference in ventilation because there was no difference in ventilation between LBP groups, only ventilation conditions. Since the soft tissue damage that might be associated with a previous injury (and its concordant potential energy deficit) was not represented in the model we must assume that this contribution was equal among participants. Since sources of stability are limited to potential energy from passive or active tissues, our interpretation is that the nLBP group had a sufficient margin of safety, probably provided by passive tissues, so that the muscles in conflict between ventilation and stability could be dedicated to the former or the latter, hence there was less often a correlation between either, ventilation and stability or, muscle activity and stability, in this group. Conversely, the yLBP group did not have the safety margin provided by passive tissue so that the musculature, particularly RA, in addition to assisting ventilation, also had to stabilize the spine leading to a correlation pattern between ventilation and stability. More muscle activity is not a requirement for greater stability, rather, as our results seem to indicate, muscle activity, appropriately synchronized with stability requirements, is necessary.

The muscles whose activity profiles were most linked to the maintenance of stability, during this task were the abdominal muscles, as determined by their statistical effects as a covariates. While RA also correlated to ventilation in the ambient air trials, in CO_2 trials yLBP subjects appeared to make subtle use of RA, at very low activity levels, to maintain stability (an effect compounded by the largest moment arm of any muscle in the model). This is illustrated by two factors (1) correlation of rectus abdominis to stability index increased in the yLBP while correlation of rectus abdominis to ventilation decreased in yLBP subjects and (2) EMG RMS differences exist between ambient air trials and CO_2 trials for both RA and EO: for both RA and EO the mean EMG RMS was greater in yLBP and greater in CO_2 trials. In addition to this, RA is associated with trial type and LBP having an interaction effect on stability index. So in yLBP subjects, when exposed to CO_2, there seems to be a greater dissociation between RA and ventilation but a better association between stability and RA. RA is rarely discussed in clinically oriented papers on stability. Anatomical and mechanical interpretation (McGill et al., 1996) shows that RA provides the anterior anchor for internal and external oblique along with transverse abdominis, suggesting that its activity has an influence on the mechanics of all these muscles. In addition, its distance from the spine makes RA the dominant flexor moment generator, magnifying its effect on stability. Of course these observations only pertain to this task, as recent work has shown that the important stabilizing muscles change as a function of the external load (Kavcic et al., 2004b).

These data suggest that changes in both passive tissue and muscle activity are linked with changing stability requirements. While this is not unexpected, it is plausible that, in the face of a more demanding task, muscle activity would compensate for damaged passive tissue and the associated loss in stiffness. This would imply that people who cannot react by either increasing muscle recruitment, recruiting new muscles or appropriately correlating muscle activity with stability demands, are at greater risk of (re)injury through instability. Conversely, it would also imply that training injured individuals for appropriate muscle recruitment strategies could reduce the risk of instability. For example, as the obliques are more involved in heavier ventilation (Abraham et al., 2002), RA can, and seems to, make up some of the difference as illustrated by an RMS average difference for EO and RA between ventilation types and a correlation to stability difference between LBP groups. The fact remains that the difference is statistically significant, consistent and in an unexpected direction. The increase in stability may or may not be functionally relevant, however, we feel that these results may shed some light on stabilizing behaviour and potential in those with a history of low back pain. In fact, stability may come from several different sources. For example, in a particular individual, passive tissue damage limits the amount of stability that those tissues can provide. Muscle stiffness could make up that difference but, in a moment of inattention, that stiffness could be lost and stability could suffer in a functional way. Instantaneous measures of stability account for this effect. We can only say that yes, the LBP group seems to overcompensate.

In a flexed posture, such as our subjects adopted, passive tissue stiffness and stability both increase (Granata and Wilson, 2001). There are two consequences to this. The first is that greater flexion strains the passive tissues more but also increases stiffness (Panjabi, 1992), the second is that greater flexion requires greater co-contraction which also increases stiffness (Crisco et al., 1992; Stokes et al., 2002). It has been shown that a higher compressive load in itself increases stiffness in the osteo-ligamentous spine (Gardner-Morse and Stokes, 2001; Janevic et al., 1991). In many cases, the oscillating activity levels of the abdominals could be offset by oscillating and drifting spine kinematics. For example, drifting into greater flexion would increase compression through greater extensor activation and balance against lower abdominal activity and even against lower stability. Thus in light loading task where the perceived threat is low and reliance on passive tissue is high, the risk of injury would increase if there is no appropriate co-activation response. The benefit of co-activation, especially at light loads, becomes evident and has been well documented (Gardner-Morse and Stokes, 2001; Granata and Orishimo, 2001). In fact, while differences in compression between groups did not reach significance, in this study, it should be noted that the cost–benefit ratio (compression to stability) reported by Granata could account for this since stability difference did reach significance while compression did not.

Several limitations should be recognized in the application of these results. First the model is based on the 50th percentile male and was not scaled to accommodate the variability in the subject population. However, the varying stress levels and optimization procedure "tunes" the model so that small anatomical variance could be accommodated. While it should be noted that the LBP group was 15% heavier that the control group, subjects were only compared to themselves such that a scaling bias would be factored out. While a one year statute was used to separate yLBP from nLBP, some who were in the uninjured category had in fact suffered a previous injury which might have resulted in lingering motor deficits, blurring the differences between the groups. Finally, a mass of twenty-two kilos was used for all subjects. Obviously, this represents a different percentage of individual strength limits.

5. Conclusion

Those with a history of LBP actually increased stability when exposed to a condition requiring increased lung ventilation. They appeared to stiffen the torso muscles which stabilize the spine partially via additional compression to the column. This has implications with regard to training injured individuals to compensate for reduced spine stability. Interestingly, the phasic relationships between stability and muscle activity patterns were seemed particularly important in maintaining or increasing stability. We speculate that passive tissue damage, imposes additional stability requirement on yLBP subjects and that this is compensated for with additional and synchronized activity of RA and EO, in phase with stability requirements rather than ventilation requirements.

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