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# How the inherent stiffness of the in vivo human trunk varies with changing magnitudes of muscular activation

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**Abstract**

*Background.* The abdominal muscles provide stiffness to the torso in a manner that is not well understood. Their unique anatomical arrangement may modify their stiffening ability with respect to the more commonly studied long strap-like muscles of the limbs. The purpose of this study was to examine stiffness inherent to the trunk, as modified by different torso, and in particular, abdominal muscle activation levels.

*Methods.* Nine healthy male participants were secured in a ‘‘frictionless’’ apparatus and subjected to applied bending moments about either the flexion/extension or lateral bend axes. Abdominal muscle activation levels were modified through biofeedback from the right external oblique muscle. Moment–angle curves were generated and characterized by an exponential function for each of flexion, extension, and right-side lateral bend, at each of four abdominal muscle activation target level conditions.

*Findings.* Stiffness measured in extension increased in a linear fashion throughout the range of motion and increased with each successive rise in abdominal activation. Stiffness in flexion and lateral bend increased in an exponential fashion over the range of motion. In flexion and lateral bend, stiffness increased with each successive rise in abdominal activation from zero to approximately 40% and 60% of the range of motion, respectively. After these points, stiffness at the highest levels of activation displayed a ‘‘yielding’’ phenomenon whereby the torso stiffness dropped below that characterized at lower levels of activation.

*Interpretation.* Increasing torso muscle co-activation leads to a rise in trunk stiffness over postures most commonly adopted by individuals through daily activities (neutral to approximately 40% of maximum range of motion). However, towards the end range of motion in both flexion and lateral bend, individuals became less stiff at the maximum abdominal muscle co-activation levels. The source and mechanism of this apparent yielding are not fully understood; future work will be directed toward elucidating the cause.


*Keywords:* Abdominal muscles; Abdominal wall; Spine; Torso

## 1. Introduction

The torso musculature is quite unique in its anatomical arrangement. In particular, the abdominal wall muscles (external and internal obliques, transverse abdominis) overlay each other in a sheet-like formation and act through attachments to the abdominal and thoraco-lumbo-dorsal fascias to create a hydraulically pressurized abdomen. These abdominal muscles, when activated, create a stiffened wall to provide stability and structural integrity to the spinal column (Farfan, 1973; Tesh et al., 1987; Cholewicki et al., 1999).

Muscle mechanics theory tells us that muscle tissue, while creating force, also provides stiffness about a joint that is at least partially dependent on the inherent spring-like stiffness of the muscle itself. Its stiffness is a combination of active components, namely myosin cross-bridge attachments, the numbers of which are dependent upon activation level and type of contraction, and passive components, namely the connective tissue network running throughout the muscle and tendon complex (Ford et al., 1981; Rack and Westbury, 1984; Lieber et al., 1992; Gajdosik, 2001). Moreover, muscle reflexes further modulate

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stiffness about a joint by reacting to a perturbation to either increase contraction to counteract motion, or to decrease contraction so as to not accentuate the motion (Nichols and Houk, 1976; Hoffer and Andreassen, 1981; Franklin and Granata, 2007). Most of what we know about muscle stiffness and its effect on surrounding joints has been obtained from studies of the long strap-like muscles of the limbs. The abdominal wall muscles, however, may not be expected to stiffen the joints of the spine in an entirely similar manner given their distinctive architecture. In fact their ability to stiffen may be enhanced through a hydraulic mechanism, modifying intra-abdominal pressure and transferring hoop stresses around the torso (Farfan, 1973; Cresswell and Thorstensson, 1989; McGill and Norman, 1993).

A number of studies have dealt with determining the effect of altering trunk muscle activation levels on trunk stiffness and/or stability by utilizing rapid perturbation paradigms (e.g. Krajcarski et al., 1999; Chiang and Potvin, 2001; Gardner-Morse and Stokes, 2001; Andersen et al., 2004; Moorhouse and Granata, 2005). In this way, these studies have captured the combined stiffness of all active, passive, and reflexive components acting within the spinal system. The consensus reached from this body of work has been that increasing muscle activation through an increased challenge imparted to the system leads to a stiffer system. More recently, Vera-Garcia et al. (2006) demonstrated that consciously increasing trunk muscle co-activation through abdominal brace techniques improved trunk stiffness in preparation for a sudden load. However, other studies have shown that attempting to consciously alter trunk muscle co-activation might constitute a non-optimal motor scheme and result in a drop in stability in more demanding situations (Brown et al., 2006).

Previous work has attempted to isolate and determine the passive, or inherent, stiffness of the in vivo trunk in each of the three anatomical planes of motion (McGill et al., 1994) and after time-varying alterations (Beach et al., 2005; Parkinson et al., 2004). To date, no study has attempted to quantify the trunk stiffness inherent, in the absence of reflexive mechanisms, at varying levels of trunk muscle activation. This may elucidate the role of torso muscle activation on the hydraulic stiffening mechanisms discussed above. Therefore, the purpose of this study was to examine trunk stiffness related to torso, and in particular abdominal, muscle activation levels, while minimizing the effect of muscle reflexes. Further, the goal was to determine the effect of increasing muscle stiffness on global trunk stiffness in each of the flexion, extension, and lateral bend directions.

## 2. Methods

### 2.1. Participants

Nine healthy male individuals volunteered from the University population (mean(SD): age 23.9(2.8) years; height 1.81(0.05) m; mass 79.0(7.1 kg)). All signed consent forms approved by the University Office of Research Ethics.

### 2.2. Data collection

Participants were secured at the hips, knees and ankles on a solid lower body platform. Each participant's upper body was secured to a cradle with a plexi-glass bottom surface, about their upper arms, torso and shoulders. The upper body cradle was free to glide overtop of a similar plexi-glass surface with precision nylon ball bearings between the two structures (Fig. 1). This jig minimizes measurable friction and allows trunk movement about either the flexion–extension or lateral bend axis, depending upon how the participant is secured. Participants lay on their right side for the flexion–extension trials, and on their back for the lateral bend trials. Their torsos were supported in each position to ensure that participants adopted and maintained a non-deviated spine posture throughout the testing. Participants were then instructed to maintain one of four torso activation patterns: relaxed (minimal activation); activate biofeedback site to approximately 5% maximum voluntary contraction (MVC) (light brace); activate biofeedback site to approximately 10% MVC (moderate brace); activate biofeedback site to approximately 15% MVC (heavy brace). Participants were instructed to tighten their abdominal muscles isometrically in order to achieve the desired brace levels. The MVCs were obtained in one of two positions: (1) a modified sit-up position in which participants isometrically attempted to produce trunk flexion, side bend and twist motions against resistance; (2) a reverse curl-up in which individuals lied supine with their hips and knees flexed to 90° while isometrically attempting to pull their thighs towards their chest, and in each of the right and left twist directions against resistance.

Once each participant had achieved their target activation level during each trial, the experimenter pulled a cable such that the upper body rotated in the desired direction. For flexion trials, the participant was pulled into flexion; for extension trials, the participant was pulled into extension; for lateral bend trials, the participant was pulled into right-side lateral bend. Participants were pulled at a relatively slow velocity (mean(SD) (degrees/s) = 5.0(2.9) flexion; 3.9(2.5) extension; 6.1(3.4) lateral bend), until a point was reached at which the experimenter could no longer effectively rotate the participant about the lumbar spine. The direction of pull of the cable with respect to the upper body cradle always remained constant; perpendicular to the upper body cradle. Once the motion began participants were no longer able to utilize the visual biofeedback to maintain their activation level; they instead were instructed to maintain the "feel" of the abdominal brace level throughout the movement. However, electromyography (EMG) was recorded throughout the trials and examined post-hoc to ensure that EMG remained near the targeted levels. Three trials of each activation condition

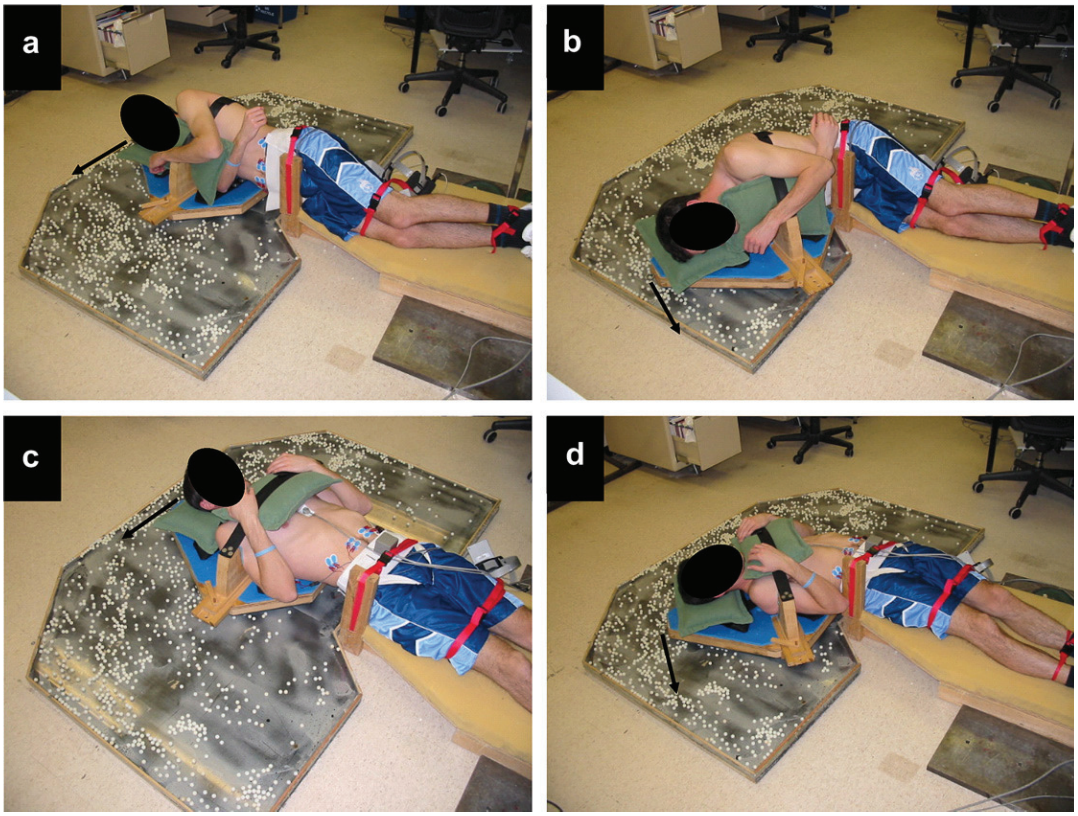


Fig. 1. Experimental set-up in the neutral position (a: flexion/extension; c: lateral bend) and at end RoM (b: flexion; d: lateral bend). Arrows indicate the direction of the applied force (perpendicular to the upper body cradle).

were conducted in a randomly assigned order for each participant.

### 2.3. Instrumentation

Fourteen channels of EMG were collected from the following muscles bilaterally: rectus abdominis (RA), external oblique (EO), internal oblique (IO), latissimus dorsi (LD), and three levels of the erector spinae (ES-T9, ES-L3 and ES-L5). Blue Sensor bi-polar Ag–AgCl electrodes (Ambu A/S, Denmark, intra-electrode distance of 2.5 cm) were placed over the muscle belly of each muscle in line with the direction of muscle fibres. Signals were amplified ($\pm$2.5 V; AMT-8, Bortec, Calgary, Canada; bandwidth 10–1000 Hz, common mode rejection ratio (CMRR) = 115 db at 60 Hz, input impedance = 10 G$\Omega$) captured digitally at 2048 Hz, low-pass filtered at 500 Hz, rectified and low-pass filtered at 2.5 Hz (single pass 2nd order) and normalized to the maximum voltage produced during isometric maximum voluntary contraction (MVC) trials to produce a linear envelope.

An EMG biofeedback device (MyoTrac, Thought Technology Ltd., Montreal, Canada) was placed in line with the right EO electrode site to allow participants to visually monitor muscle activity at this level.

Three-dimensional trunk motion was recorded using an electromagnetic tracking system (Isotrak, Polhemus, Colchester, VT, USA) with the source secured over the sacrum and the sensor over T12 for the flexion/extension trials, and the source over the lower abdomen at a level slightly below the ASIS and the sensor over the xiphoid process for the lateral bend trials. The trunk motion data was sampled digitally at 32 Hz and dual-pass filtered (effective 4th order 3 Hz low-pass Butterworth).

The moments applied to the torso were recorded by the product of the force applied perpendicular to the distal end of the upper body cradle and the moment arm from the location of the applied force to the level of L4/L5. Force was recorded with a force transducer (Transducer Techniques Inc., Temecula, CA, USA) and digitally sampled at 2048 Hz. Force signals were dual-pass filtered (effective 4th order 3 Hz low-pass Butterworth). Both the linear enveloped EMG and force signals were downsampled to 32 Hz to match the trunk motion data.

### 2.4. Moment–angle curves

The applied moment and corresponding trunk angle were windowed for each trial and normalized in time to ensure equal trial length across all trials and participants. Trunk angles were normalized as a percentage of the maximum range of motion (RoM) that participants were able to obtain in trials conducted from an upright standing position.

Data were combined across trials and subjects for each muscle activation/brace level for each of the flexion, extension, and lateral bend directions. Exponential curve fits of the following form were performed for each brace level/direction combination:

$$M = \lambda e^{\delta\phi}$$

where $M$ is the applied moment (N m); $\lambda$ and $\delta$ are the curve fitting coefficients and $\phi$ is the trunk angle as a percent of the standing max RoM.

This equation was differentiated once with respect to $\phi$ to obtain a measure of trunk angular stiffness:

$$K = \lambda\delta e^{\delta\phi}$$

where $K$ is the angular stiffness (N m/%RoM).

Additionally, the applied moment required to initiate trunk motion, the peak applied moment, and the maximum trunk angular displacement were all recorded for each trial. The normalized EMG activation averaged over each of the 250 ms prior to the initiation of the applied moment, as well as the 250 ms prior to the end of movement, was quantified and averaged across the right- and left-side muscles. For a comparison in activation levels between each of the two 250 ms periods, right- and left-side muscles were kept separate for the lateral bend condition.

### 2.5. Statistical analysis

Each of the dependent variables was averaged within each subject for each condition.

Repeated Measures 1-way (four muscle activation levels) ANOVAs were conducted for each of the following independent variables: average EMG activation prior to applied moment initiation for seven muscle sites; the applied moment required to initiate trunk motion; the peak applied moment; and the maximum trunk angular displacement. The effect of time on muscle activation levels, and possible interactions with brace levels, were evaluated using a Repeated Measures 2-way (muscle activation level and EMG pre versus final 250 ms of movement) analysis of variance (ANOVA). Tukey's honestly significantly different (HSD) tests were run in cases where a significant effect ($P < 0.05$) was determined by the ANOVA.

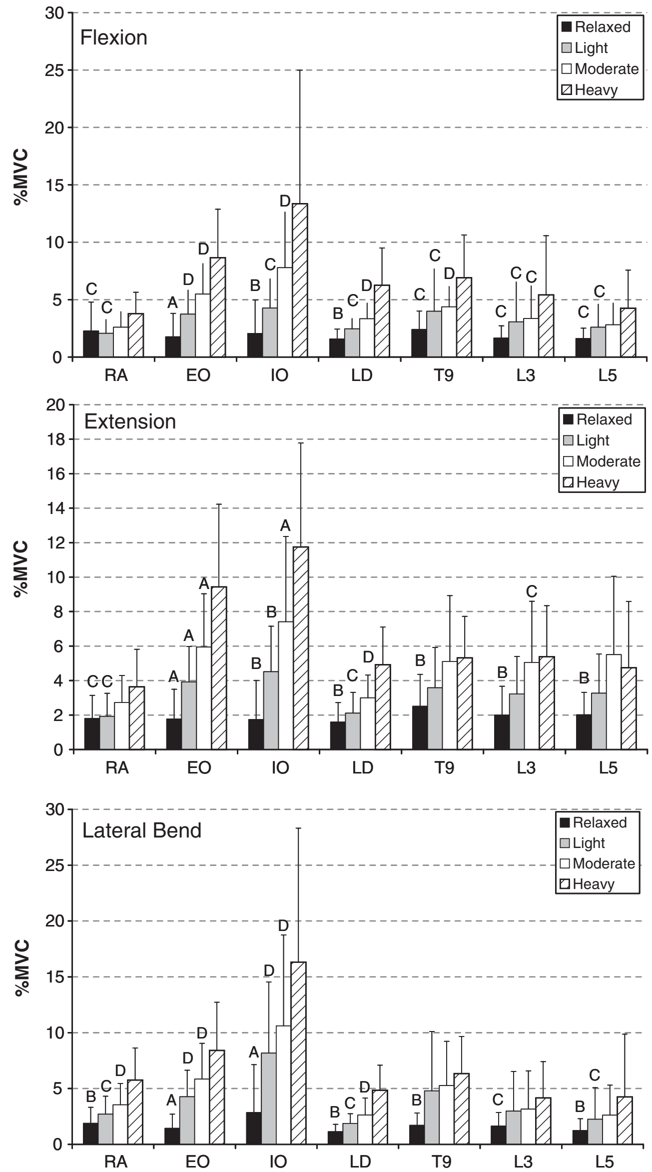


Fig. 2. EMG averages (across all trials and bilaterally across right and left sides) and standard deviations for the 250 ms prior to the initiation of the applied moment for each brace level in each of the flexion, extension and lateral bend conditions. Significance ($P < 0.05$) is as follows: $A$ = different from all other levels; $B$ = different from the moderate and heavy levels; $C$ = different from the heavy level; $D$ = different from the relaxed and heavy levels.

## 3. Results

### 3.1. EMG

Average muscle activation levels, quantified prior to the initiation of movement, increased between each of the relaxed, light, moderate, and heavy abdominal brace levels for every muscle except the RA between the relaxed and light brace levels in the flexion condition and the ES-L5 between the moderate and heavy brace levels in the extension condition (Fig. 2). Statistically significant ($P < 0.05$) differences between levels are indicated in Fig. 2.

A number of statistically significant differences, consistent across all brace levels, were found for average muscle activation levels prior to versus at the end of movement. Those that increased activation from initiation to the end of the movement were: in extension ES-L5 (3.9–6.0%MVC); in lateral bend right RA (4.3–7.2%MVC), right EO (6.2–10.4%MVC), right ES-T9 (3.7–5.3%MVC) and left ES-T9 (5.3–7.7%MVC). Those that decreased activation from initiation to the end of the movement were: in flexion ES-L5 (2.8–1.8%MVC); in lateral bend left ES-T5 (2.1–1.2%MVC). The more interesting muscles were those that showed an interaction between time and brace level (in flexion LD and EO; in extension LD; in lateral bend both right and left LD). The LD in flexion and both LDs in lateral bend increased activation towards the end of movement in all brace conditions, with greater differences between the two time periods for each successive increase in brace level. In extension the LD showed a lower activation level at the end of movement in the relaxed condition but a higher level at the end of movement in each of the three brace conditions. The EO in the flexion trials was the only muscle to display a decrease in activation level at the end of movement that was greater with each successive magnitude of brace level. Despite these documented changes in activation level, a very similar pattern of increased torso muscle activation between each of the abdominal brace levels existed for all muscles over the last 250 ms of movement as did prior to the initiation of movement.

### 3.2. Stiffness curves

The light brace flexion moment–angle data, combined across all trials, is displayed as an example in Fig. 3. Least squares best fit stiffness curves are shown, encompassing zero to 100% of the maximum standing RoM, for each of the flexion, extension, and lateral bend directions (Fig. 4). Stiffness increased exponentially at each muscle activation level in both flexion and lateral bend. In flexion, from zero to approximately 40% RoM, stiffness increased with each level of abdominal brace; in lateral bend, this trend existed from zero to approximately 60% RoM. At the end RoM in flexion, individuals were stiffest when employing a light muscle activation level, followed by relaxed, with heavy and moderate activation levels displaying the lowest stiffness levels. In lateral bend, at the end RoM, individuals were stiffest when employing a moderate

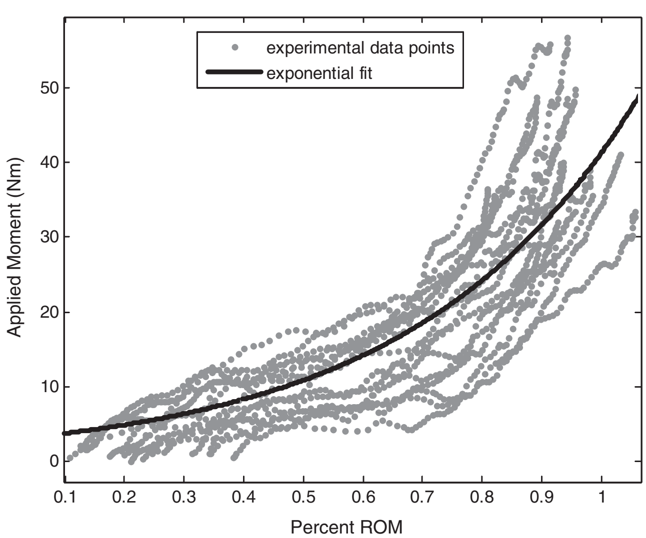


Fig. 3. Scatterplot of moment–angle data points for all trials and participants within the light brace flexion condition. Thick line indicates exponential line of best fit.

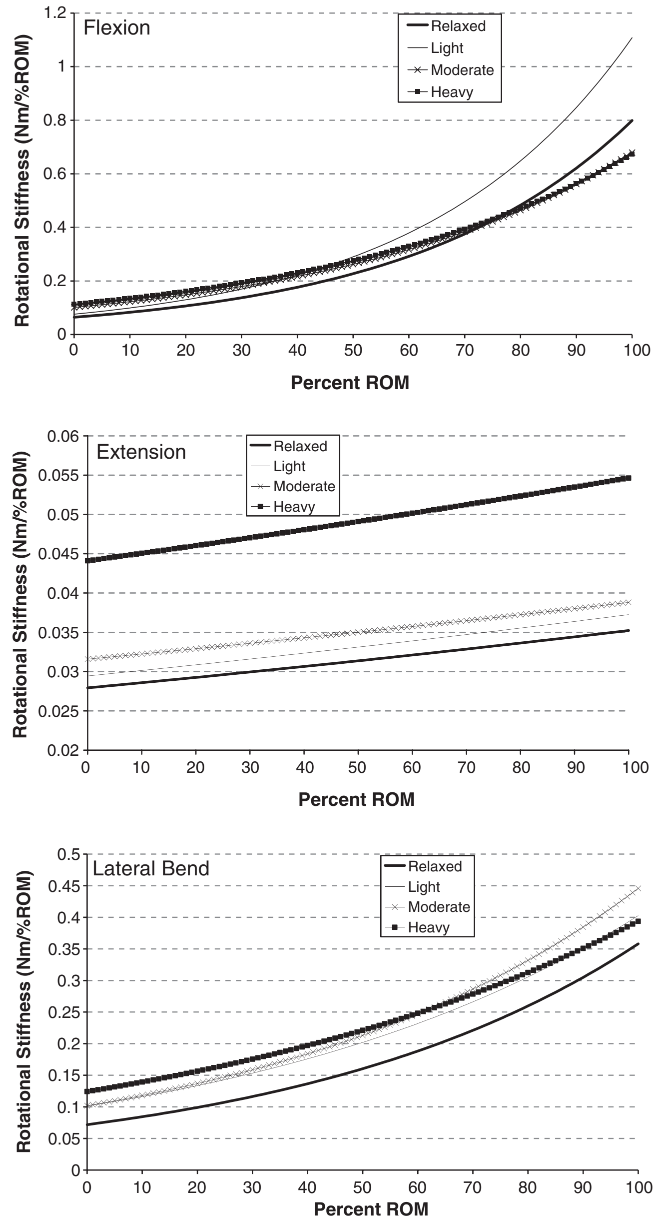


Fig. 4. Stiffness (N m/%RoM) determined from the first derivative of the moment–angle curve fits ($K = \lambda\delta e^{\delta\phi}$) across the RoM in each of the flexion, extension and lateral bend directions.

level of activation, followed by the light and heavy levels, with relaxed displaying the lowest stiffness levels. Best-fit coefficients are displayed in Table 1.

Stiffness in extension showed an increasing linear trend with increasing RoM for each of the muscle activation levels. Stiffness increased with each successive increase in trunk muscle activation.

### 3.3. Moment–angle characteristics

A higher applied moment was required to initiate movement in the heavy brace as compared to the relaxed condition in each of the flexion ($P = 0.028$), extension

Table 1
Best fit coefficients and root-mean-square (RMS) error (N m) for equation 1 ($M = \lambda e^{\delta\phi}$) for the relaxed and each of the three different brace levels in each of flexion, extension, and lateral bend

| | Flexion | | | | Extension | | | | Lateral bend | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | Relaxed | Light | Moderate | Heavy | Relaxed | Light | Moderate | Heavy | Relaxed | Light | Moderate | Heavy |
| $\lambda$ | 2.565 | 2.831 | 5.531 | 6.354 | 12.030 | 12.520 | 15.350 | 20.580 | 4.474 | 7.353 | 6.913 | 10.760 |
| $\delta$ | 0.0252 | 0.0268 | 0.0190 | 0.0178 | 0.00232 | 0.00235 | 0.00206 | 0.00214 | 0.0161 | 0.0138 | 0.0148 | 0.0115 |
| RMS | 5.17 | 7.31 | 9.21 | 10.23 | 8.04 | 7.11 | 8.31 | 11.77 | 8.55 | 10.82 | 9.86 | 13.24 |

($P = 0.025$), and lateral bend ($P = 0.025$) directions (Fig. 5). Additionally, the peak applied moments (corresponding to the point at which the experimenter could no longer rotate the participant) were significantly higher in extension ($P = 0.004$) in the heavy brace as compared to the relaxed condition, and in lateral bend ($P = 0.043$) in the heavy brace as compared to the light brace condition (Fig. 6). Finally, the maximum trunk angular displacement was significantly greater in the relaxed as compared to the heavy brace condition in the flexion direction ($P = 0.0245$) (Fig. 7).

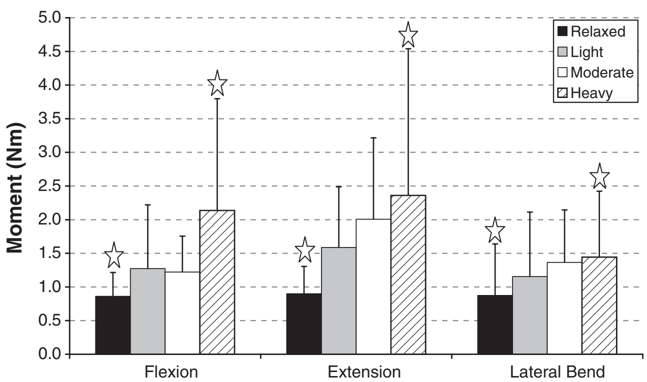


Fig. 5. Average (SD) moment required to initiate bend about each axis in each of the relaxed and light, moderate and heavy brace conditions. Conditions, within each bend direction, which are significantly different ($P < 0.05$) from one another are indicated with stars.

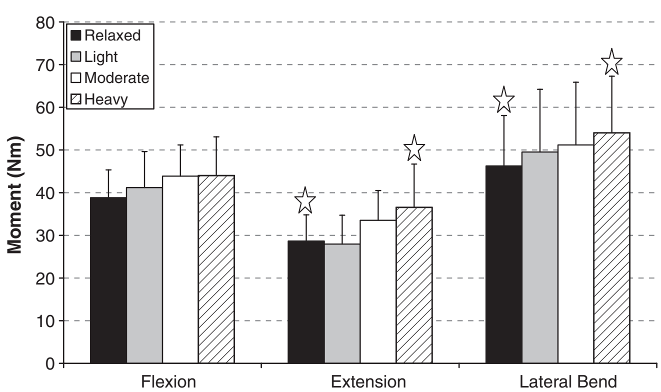


Fig. 6. Average (SD) peak moment, corresponding to the end RoM, about each axis in each of the relaxed and light, moderate and heavy brace conditions. Conditions, within each bend direction, which are significantly different ($P < 0.05$) from one another are indicated with stars.

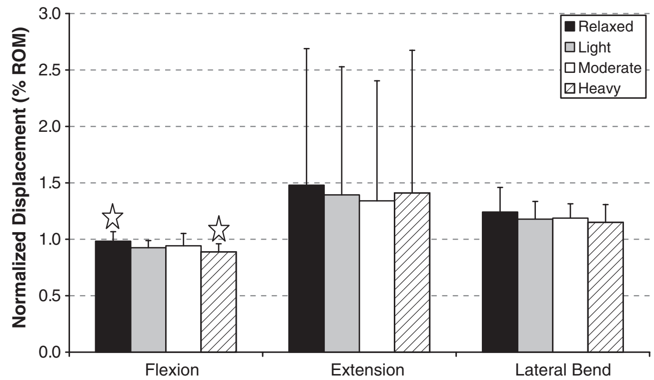


Fig. 7. Average (SD) maximum trunk displacement, normalized to the maximum attained in standing RoM tests, about each axis for each of the relaxed and light, moderate and heavy brace conditions. Conditions, within each bend direction, which are significantly different ($P < 0.05$) from one another are indicated with stars.

## 4. Discussion

The primary purpose of this study was to determine the amount of torso stiffness inherent to the trunk musculature, and in particular the abdominal musculature, at different levels of activation. It was found that stiffness increased with each successive increase in muscle activation level across the entire RoM in a linear fashion in extension, and in a non-linear fashion with stiffness increasing at a greater rate at higher angles of rotation, through the low to mid RoM (neutral to approximately 40% of maximum) in each of flexion and lateral bend (Fig. 4).

Muscle activation levels were manipulated through the use of abdominal bracing techniques. In this technique, individuals focus on isometrically tightening, or increasing activation levels, of the abdominal wall musculature. The isometric nature of this task induces opposing muscle groups, primarily the trunk extensors, to concomitantly increase activation (Fig. 2). In addition, contraction of the abdominal wall stiffens posterior components of the spine via interaction with the lumbo-dorsal fascia (Tesh et al., 1987), and creates associated increases in intra-abdominal pressure (Cholewicki et al., 1999; Essendrop et al., 2002; Hodges et al., 2005). In the current study, varying levels of bracing were achieved through the use of visual biofeedback from the right external oblique muscle site. Therefore, the largest increases in activation between each of the brace levels were seen in the external and internal oblique muscles. Highest activation levels reached approximately 16% in the internal oblique in the lateral bend conditions, and 12–13% in the internal oblique in the flexion and extension conditions. The greatest activation changes between adjacent brace levels tended to occur between the moderate and heavy braces, and the smallest between the light and moderate brace levels. For the majority of the participants, the heavy brace level represented the maximum isometric abdominal contraction that they could achieve in the test position. They therefore were able to somewhat remove focus from the biofeedback and tend focus to attaining maximal contraction in these trials.

It was initially hypothesized that for each direction of movement, stiffness would increase along with successive increases in muscle activation. This was confirmed throughout the extension RoM, and in the flexion and lateral bend directions for the first 40–60% of RoM. For reasons that are not fully understood, there appeared to be a "yielding" phenomenon occurring with higher levels of activation as end range of flexion and lateral bend were approached. There are two possible explanations for this finding: (1) Activation of the abdominal wall muscles creates a balloon-like structure of the abdomen. Increasing activation raises the tension and creates a stiffer balloon. As bending occurs, the balloon eventually folds upon itself, thus yielding its increasing resistance to bend; the stiffer the original state of the balloon, the greater the load acting upon it and thus the greater the yielding effect; (2) The light and moderate brace levels were much easier to attain for the participants, and it is therefore plausible that individuals had difficulty in controlling the more difficult brace levels during the mid to upper ranges of the RoM. Indeed, the activation levels of certain muscles changed over the course of the movement, displaying different levels over the last 250 ms of movement as compared to the period prior to the initiation of movement. These changes were, however, counter to what one might expect to create the apparent "yielding" effect seen here; the muscles either changed consistently across the different brace levels or showed greater increases in activation at the higher levels of abdominal bracing. Still, it has been shown previously that increasing activation in isolated muscles can create an imbalance in torso stiffness (Brown and McGill, 2005; Brown et al., 2006). This idea is consistent with work showing that consciously increasing activation in the torso musculature can potentially degrade postural control (Reeves et al., 2006) and elevate motor control difficulty, thereby compromising torso stiffness in more challenging situations (Brown et al., 2006).

A number of factors contributed to the trunk stiffness examined in the current study. During rapid length changes, muscles display a "short-range" stiffness that is proportional to the number of strongly attached cross-bridges to produce contraction (Joyce and Rack, 1969; Ford et al., 1981; Ettema and Huijing, 1994). This stiffness lasts only for very small length changes, until cross-bridge bonds break, and is most pronounced at high velocities (Rack and Westbury, 1984; Mutungi and Ranatunga, 1996). Due to the slow velocity, long-range nature of the stretches in the current study, it is unlikely that the muscles displayed the full potential stiffness residing in the cross-bridges. Some additional stiffness inherent to the muscle may reside in the reorganizing of the intra-muscular and extramuscular connective tissues that occurs with contraction (Monti et al., 1999; Meijer et al., 2006).

Tissues directly unrelated to muscle activation provide additional stiffness to the trunk, especially as end RoM is approached. Ligaments and intervertebral discs (Adams et al., 1980), buckled abdominal contents, and bony geometry all provide varying amounts of stiffness towards end RoM in each of the three motion directions. Because these factors are a function of spine posture and tissue length, their stiffness contributions would be the same for each level of muscle activation. Furthermore, an increase in intra-abdominal pressure coincides with increased abdominal muscle activation (Cholewicki et al., 1999; Essendrop et al., 2002), which also results in increased spine stiffness (Cholewicki et al., 1999; Essendrop et al., 2002; Hodges et al., 2005).

A limitation of this study that may have additionally confounded the end RoM data is the structure and shape of the passive motion jig itself. Care was taken when securing participants on the lower and upper body cradles to allow freedom of movement through as much of the RoM as possible. However, towards the very end of movement in flexion and lateral bend, participants occasionally became partially obstructed by contact between the two cradles; this was then considered the end point so as not to affect the stiffness estimates. Individuals for whom this was the case all stated that they felt like they were at or very near their true end RoM when the movement ended. A second limitation is that no separation was made between viscous and elastic resistive forces; credit for all resistance to the applied moment was given to the stiffness of the system. Thus, the stiffness curves in the current study represent a simplified effective stiffness of the trunk. Finally, nine healthy males participated in the current study. A larger and more diverse sample population might help to shed light onto the cause of the relatively unexpected findings regarding the potential yielding of trunk stiffness at the highest activation levels towards the end RoM.

## 5. Conclusions

The ability of increasing torso, and in particular abdominal, muscle activation to increase trunk stiffness is partially dependent upon trunk posture. In extension, spine stiffness increased with successive increases in muscle activation throughout the RoM. Similarly, in trunk postures most commonly adopted by individuals through daily activities (neutral to approximately 40% of maximum RoM) spine stiffness increased in the flexion and lateral bend directions as muscle activation increased. However, towards the end RoM in both flexion and lateral bend, individuals became less stiff at the maximum abdominal muscle co-activation levels. The source or mechanism of this apparent yielding phenomenon is not yet clear; future work will be directed to uncover the cause.

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