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Lower Torso Muscle Activation Patterns for High-Magnitude Static Exertions

Gender Differences and the Effects of Twisting

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Study Design. Surface electromyographic signals were collected from 14 lower torso muscles while participants resisted high-magnitude static trunk moments applied in a variety of directions.

Objectives. To obtain a description of muscle activations in response to large moment magnitudes and axial twisting, including levels of agonistic and antagonistic muscle cocontraction. To assess differences in lower torso muscle activation patterns associated with gender and trial repetition.

Summary of Background Data. Back pain is associated with mechanical loads in the back. Biomechanical modeling of these loads is facilitated by knowledge of typical muscle activation patterns. Previous efforts in obtaining such data have often limited their scope to low-magnitude exertions or relatively simple scenarios.

Methods. Eight male and eight female participants, matched by height and mass, performed static exertions in an apparatus that immobilized their lower body while the activation levels of seven bilateral torso muscles were measured using surface electromyography. Activation patterns were analyzed to assess differences resulting from a variety of factors.

Results. No significant differences in activation patterns were found between genders or repetitions, but moment magnitude and direction elicited substantial differential responses. Good repeatability was found between trial repetitions, as indicated by intraclass correlation coefficients (>0.65). Significant synergistic muscle coactivation, large intersubject variability (mean coefficient of variation 82.2%), and consistent levels of antagonism ranging from 10% to 30% maximum voluntary exertions were observed.

Conclusions. Individuals of different genders, but similar anthropometry, have comparable muscular reactions to complex torso loads, suggesting similar motor control strategies. Future spine models should consider that the variability in muscle recruitment patterns is larger between subjects than within subjects. High-magnitude exertions, especially those with moment loads in more than one plane, require most muscles to be active (>5%) and moderate levels of antagonism. [Key words: electromyography, muscle activation patterns, twisting, spine biomechanics, trunk muscles] **Spine 2002;27:1326–1335**

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Despite a growing body of research concerning the prevention, diagnosis, and treatment of low back pain, the disorder continues to be a costly and widespread occupational problem. Many task characteristics have been studied in order to understand their effects on the onset or aggravation of low back pain, and associations have been observed between several of these characteristics and the incidence of low back pain.[23,37,62,68] Some researchers have undertaken the considerable task of translating these task characteristics into their resultant internal spinal loading, with varying degrees of success based on the modeling approach and the type of task being modeled.[3,10,12,16,20,27,44,47,49,50,54,59,66] Closer study of these efforts reveals a number of uncertainties regarding spinal mechanics, evident in the number of unknowns that spine models must address. Analysis of muscle recruitment patterns can provide information about low back muscle forces, which are largely responsible for the forces acting on the spine and consequently essential in predicting the risk of low back injury.

The most complete description of trunk muscle activations has been compiled by Lavender and colleagues.[30,31,34–36] Their data were based on surface electromyography (sEMG) collected from participants resisting a wide range of static moments. Their studies demonstrated the presence of antagonism in lower torso muscles, the dependence of muscle activation on trunk moment direction and magnitude, and the effects of postural changes on muscle activation patterns. Several other studies have examined the effects of various loading conditions and moment magnitudes, in some cases dynamically.[14,28,29,32,33,40,45,53,55,61,67] In many of these studies, explicit hypotheses were evaluated concerning the strategies used to counteract lumbar moments using a complex set of muscle activations. However, comparisons or combinations of the data from these investigations, which could result in a more comprehensive representation of muscle activity and coordination, are difficult because of diverse experimental methods, participant pools, and experimental goals.

There are two primary limitations in most existing reports. First, muscle activities have typically been measured in response to low moment magnitudes relative to isometric back strength. For example, Lavender et al[35] applied 50 Nm in extension and flexion to participants, whereas back strength in those directions has been estimated by Kumar[25] to be near 206.5 Nm in extension and 130 Nm in flexion (male and female average). Patterns of muscle activity might differ between conditions of high and low lumbar moments. Second, consideration of moments about the horizontal plane has, with some exceptions,[13,39,56,65] been limited to whether or not there is a twisting moment in the trial[21,26,34,38,41–43,51,63] but with no systematic variation of the axial moment magnitude. This void in the literature is particularly important in the study of muscle antagonistic coactivation. If present, antagonism requires agonist muscles to exert force to counteract not only the original moment but also the antagonistic force. This increased agonist activation results in increased spine compression and shear forces, which may increase the risk of low back injury.[17]

Intrasubject variability, as measured using task repetition, has as well been addressed rarely in reported studies. Whereas intersubject variability has been clearly demonstrated, the extent of intrasubject variability for lower trunk muscles[47] has received less attention. Some have argued that small levels of intrasubject variability observed in occupational settings obviate the need for repeated sampling.[1] This argument, however, requires more quantitative support.

Gender and its influence on muscle activation patterns, has also been rarely considered in most previous investigations, with participant pools often restricted to males. Although several studies have demonstrated some degree of gender differences in muscle activation,[4,5,9,15] no such studies have been reported for the lower trunk muscles. Gender differences in trunk muscle activation are important because of the increased exposure of females to physically demanding jobs.[64] If females are more susceptible to low back injury than males, then proper accommodations for females must be made in the design of these jobs.

In the present study a database of muscle activation patterns was compiled that complements previous investigations. Variations in activation patterns were quantified between different individuals, different trial repetitions in the same individual, different genders, different loading magnitudes, and different loading directions. The following hypotheses were tested: 1) activation patterns vary more between individuals than within an individual, and 2) gender has no effect on the activation patterns. In addition, patterns of synergism and antagonistic coactivation were evaluated with respect to several characteristics of the applied loads.

■ Methods

Participants. Eight males and eight females participated in the study. All of the participants completed an informed consent procedure before beginning the experiment, the protocols for which were approved by the Virginia Tech Institutional Review Board. Individuals reporting any occurrences of disabling back pain in the past year, any current low back pain, or any back surgery were not allowed to participate. To reduce the effects of anatomic differences between the participants, males and females of similar heights ($P > 0.69$) and body masses ($P > 0.11$) were recruited. Body masses (average ± SD) for males and females were 78.7 ± 3.59 kg and 75.8 ± 3.38 kg, respectively. Heights for males and females were 171.9 ± 3.61 cm and 170.9 ± 5.75 cm, respectively. Participants' ages were 27 ± 6.93 years for males and 25 ± 2.51 for females.

Data Collection. Participants performed specified static exertions while standing with their feet on a force platform (Bertec, Columbus, OH) and separated at about shoulder width. To minimize movement of the legs and pelvis during the exertions, the knees and pelvis were restrained by straps against a rigid fixture bolted on top of the force plate, an approach described previously.[19,54] External moments were applied through a belt-type shoulder harness, similar to the methods of Lavender and colleagues,[30,31,34–36] and kept in place using skid-resistant tape between the harness and clothing. The harness had load attachment points bilaterally and at the front and rear midlines.

sEMG recordings were obtained from selected trunk muscles using bipolar (Ag/AgCl) disposable electrodes. The electrodes were placed bilaterally in standardized locations over seven muscle pairs: longissimus thoracis,[65] iliocostalis lumborum,[6] multifidus,[6] rectus abdominis,[44] latissimus dorsi,[44] internal oblique,[44] and external oblique.[22] Before placing the electrodes, the skin was abraded and cleansed with alcohol to keep electrode impedance under 40 kΩ. sEMG signals were transmitted through short (<30 cm) leads to preamplifiers (100× gain), then instrumentation amplified, band-pass filtered (30–1000 Hz), RMS converted (110 msec time constant), AD converted (512 Hz), and stored on disc. The recordings were later normalized (NEMG) against maximal values obtained from a set of maximum voluntary exertions (MVEs) performed in directions of extension, flexion, left and right lateral bending, clockwise and counterclockwise axial rotation, and attempted one-armed brachiation for each of the arms.

Independent Variables. Static moments were varied systematically about two out of three orthogonal planes in the following combinations: sagittal–frontal, sagittal–horizontal, and frontal–horizontal (Figure 1). In each of these three pairs of orthogonal planes, the moment direction was set to 0°, 45°, 90°, 135°, and 180° (Figure 1; similar to the approach of Lavender et al,[36] who used seven equally spaced orientations). Furthermore, the moment magnitude was varied as a percentage of the vector combination of MVEs for each participant, using an interpolation method to estimate biplanar strengths from the uniplanar MVEs measured at L3–L4 (Figure 2).

A full factorial combination of the previous variables results in 15 treatments per moment magnitude. Any duplicates in the matrix of moment combinations (Figure 1) were removed, yielding 11 of 15 (Figure 1) in different moment directions that were applied. Two moment magnitudes (50% and 90%) were used, resulting in 22 different moment loads per participant. Each of these 22 loads was presented twice. The resulting 44 trials were divided into two identical and completely randomized blocks to prevent any repetition of a trial from taking place before all the initial trials had been completed. After load application participants were allowed to stabilize their postures by straightening their torso with feedback provided by elastic guides. EMGs were then measured for 3 seconds, and at least 2 minutes of rest was provided between each exertion.

Data Analysis. Differences in MVE levels between females and males were tested using a multiple analysis of variance (MANOVA) followed by analysis of variance of the four MVE directions (*i.e.*, flexion, extension, lateral bending, and twisting). Another MANOVA was performed on the NEMG values

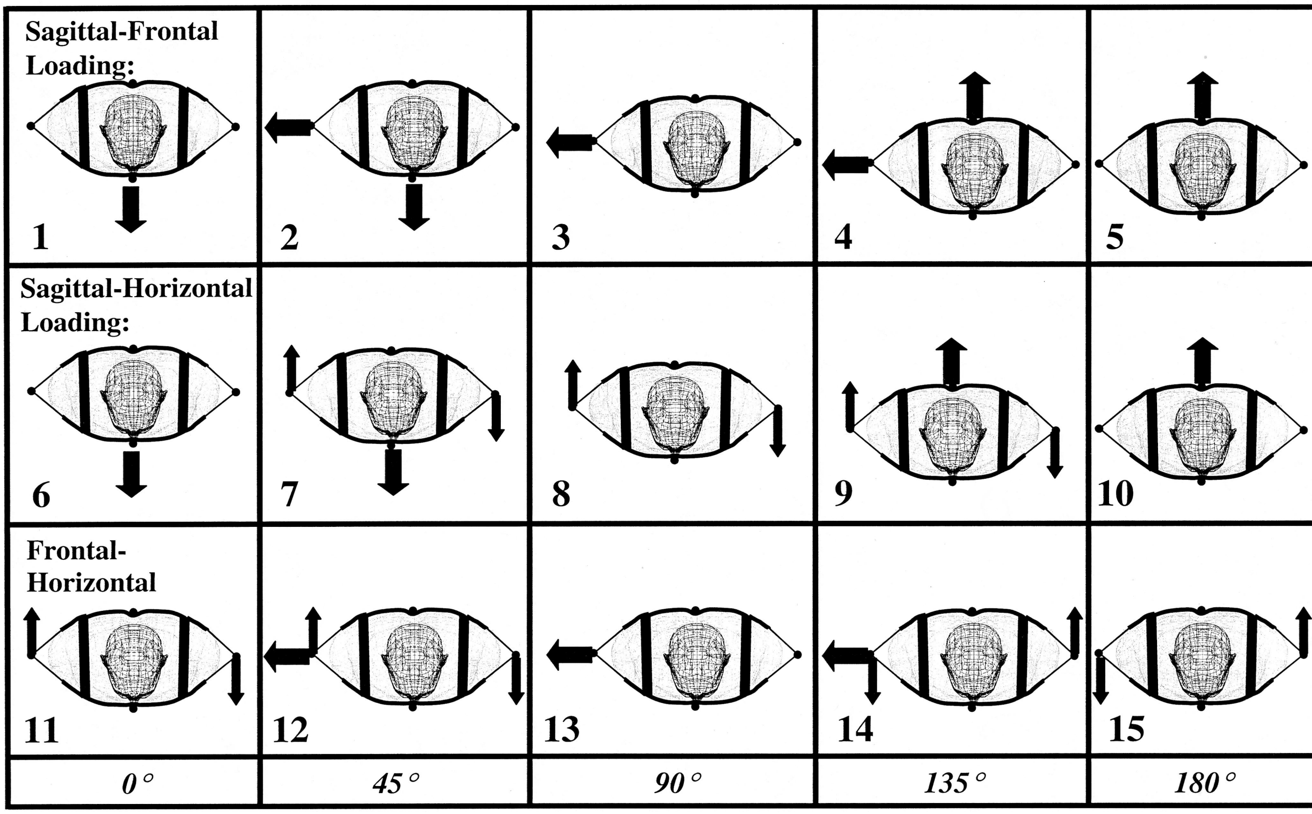

Figure 1. Moment directions for loadings across pairs of torso axes. Arrows indicate the load attachment points and load directions used to obtain the desired combination of moments. Moments in the three planes were achieved by changing the load's attachments to a harness. For the sagittal–frontal plane loading, 0° represents applied flexion, 90° represents right lateroflexion, and 180° represents applied extension. For the sagittal–horizontal plane loading, 0° represents applied flexion, 90° represents counterclockwise twisting, and 180° represents extension. For frontal–horizontal plane loading, 0° represents counterclockwise twisting, 90° represents right lateroflexion, and 180° represents clockwise twisting. The following pairs of cells are duplicates: 1 and 6 (applied flexion), 3 and 13 (applied lateral bending), 5 and 10 (applied extension), and 8 and 11 (counterclockwise twisting).

considering the following factors and all their interactions: gender, repetition, moment magnitude, and moment direction, where gender is the only between-subjects factor. Statistical power of the F tests was estimated for the nonsignificant main effects and two-way interactions in the ANOVA. Intraclass correlation coefficients (ICCs) [(2,1) case][58] were used to determine the consistency of activations between repetitions of the same trial. Coefficients of variation were used to compare intersubject and intrasubject variability. One-sided, one-sample t tests were used to determine whether the mean activity level, across subjects, was significantly greater than 5% MVE ($P \leq 0.05$), the assumed muscle activation threshold. To evaluate synergistic activation (coactivity) of the muscles, two-sample t tests were used to determine differential muscle activation between the two magnitudes at each loading direction. For these latter tests, effects with $P \leq 0.003$ (Dunn–Bonferroni procedure for multiple *a priori* comparisons yielding $\alpha = 0.05$) were considered significant.

Results

MVEs

MVEs for females and males are summarized in Table 1. Whereas gender was a significant effect in the MANOVA ($P = 0.0021$), only the twisting moments exhibited significant differences between genders when evaluated in ANOVAs ($P = 0.0085$ for twisting, $P > 0.7029$ for other exertions).

Effects of Loading Condition, Gender, and Repetition

As expected, the main effects of moment magnitude and moment direction were all significant ($P < 0.0001$). The magnitude by direction ($P < 0.0001$) interaction was also significant. Neither gender ($P = 0.6913$), nor repetition ($P = 0.2194$), nor any of their interactions with other factors ($P > 0.2166$) was significant, although the statistical power of these tests was large ($\beta > 0.80$). The overall ICC between repetitions was 0.963. ICCs were also calculated with the data grouped by muscle (Figure 3). Flexor muscles generally had larger ICCs than extensor muscles, but all ICCs indicated moderate to high repeatability (ICCs > 0.65).

Activation Patterns

Graphs of the average activation patterns per muscle and loading condition were generated for loading conditions without twisting and with twisting (Figures 4, 5, and 6, and 6) and for the 50% and 90% moment magnitudes. As suggested by the MANOVA results, differences in activation patterns could be observed between different moment directions and moment magnitudes. In general,

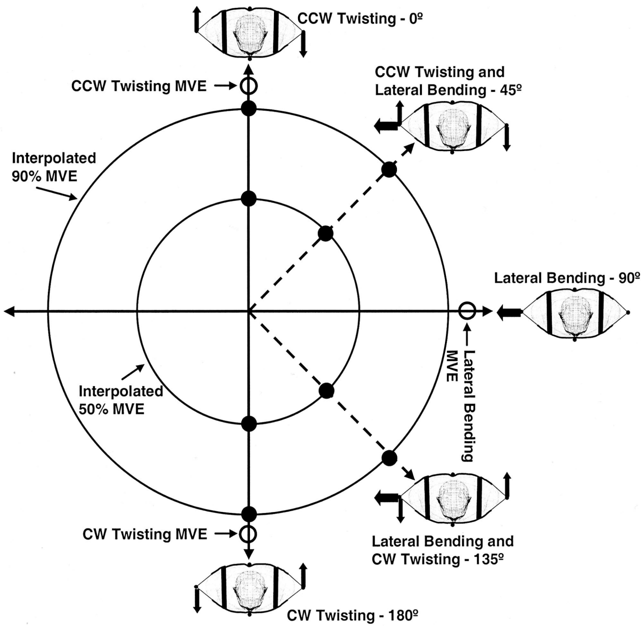

Figure 2. Frontal–horizontal plane loading across pairs of torso axes and graphical indication of the MVE interpolation method. Axes are first scaled such that the participant's uniplanar MVEs are equidistant from the origin. Concentric circles then indicate the combinations of uniplanar moments that will approximate the intended moment at any specific orientation (*e.g.*, the outer circle indicates 90% MVE). The small filled circles indicate the experimental conditions considered in this study. Experimental MVEs are indicated by small unfilled circles.

Table 1. Average MVEs for Male and Female Participants in Different Exertion Directions

	Male	Female	Average
Flexion (Nm)	197.5 (98.5)	210.6 (97.5)	204.0 (98.0)
Extension (Nm)	211.0 (104.3)	219.4 (106.0)	215.2 (105.2)
Lateral bending (Nm)	131.8 (70.6)	117.3 (59.0)	124.6 (65.1)
Twisting (Nm)	90.7 (43.1)	48.3 (22.8)*	69.5 (34.5)

* Significant difference ($P < 0.05$): male *vs.* female.

intersubject differences, as indicated by the standard deviation, were moderate across muscles, moment directions, and moment magnitudes. The largest observed standard deviations were around 40% of maximal EMG values. The average intersubject coefficient of variation, a measure of variability relative to the mean, was 82.2%, with a maximum value across all loading conditions of 204.5% (Figure 6b, right iliocostalis lumborum muscle, 90% MVE). The variance itself also showed a tendency to increase with the mean activation levels, suggesting higher levels of variability between subjects for muscles that were heavily recruited. Coefficients of variation for intrasubject variability were considerably lower, with a mean value across all individuals of 29.7% and a maximum value of 93.4%.

Most muscles appeared to be in an active state (NEMG > 5%) for all loading conditions and particularly when biplanar moments were applied (Figures 4 to 6). Furthermore, changes in moment magnitude were associated with changes in the activation levels of several, but not all, muscles monitored.

In the trials for which a direct comparison between twisting and nontwisting conditions can be made (*e.g.*, Figure 4c *vs.* Figures 4a and 5c), muscle activation resulting from a combination of loads seems to follow an "additive" pattern. If the activations of Figure 4a (pure extension) and Figure 5c (pure twist) are added, for example, the activation patterns are similar to, although slightly lower than, the ones presented in Figure 4c (extension and twist). A similar observation can be made for other trials [*e.g.*, Figure 4d (flexion and twist) *vs.* Figure 4b (pure flexion) and Figure 5c (pure twist); Figure 6a (extension and lateral bending) *versus* Figure 4a (pure extension) and Figure 6c (pure lateral bending)]. Ob-

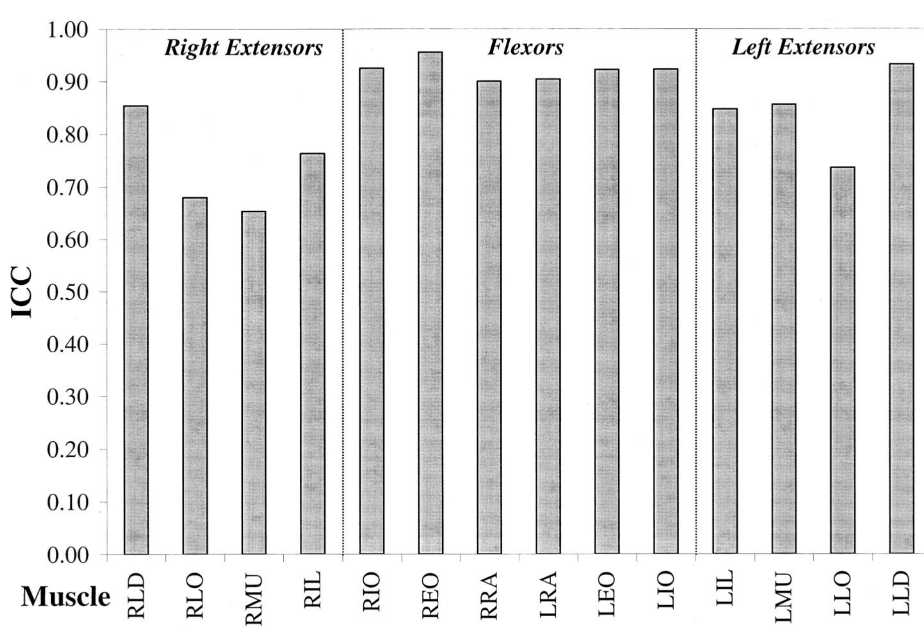

Figure 3. Intraclass correlation coefficients (ICCs) for each muscle calculated using all experimental trials. Flexors had, in general, larger ICCs than extensors.

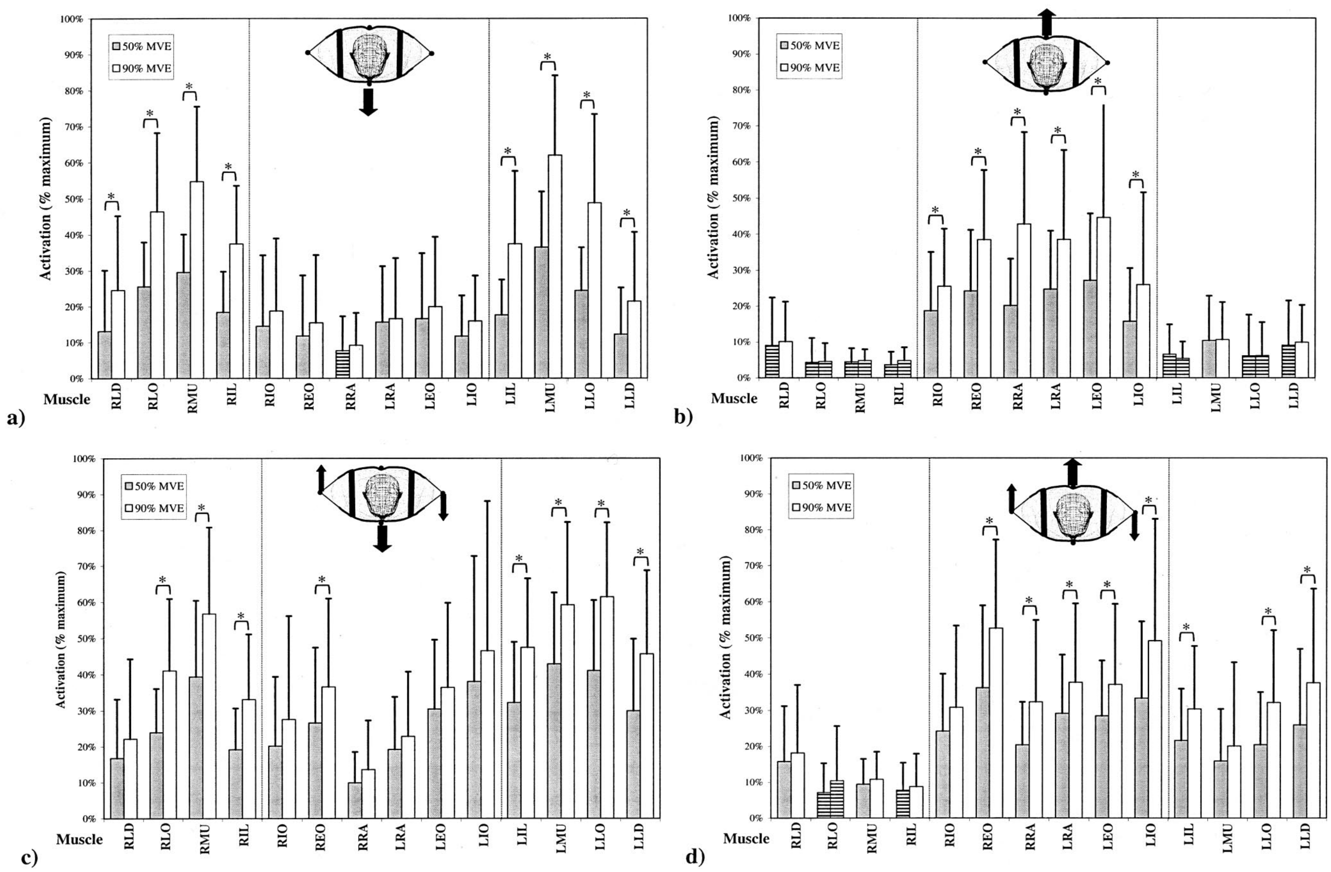

Figure 4. Normalized EMG for each muscle and moment direction under the following: **(a)** applied flexion with no twist, **(b)** applied extension with no twist, **(c)** applied flexion with counterclockwise twist, and **(d)** applied extension with counterclockwise twist. The illustration accompanying each graph indicates the loads applied to the participant. Vertical dotted lines separate, from left to right, right extensors, flexors, and left extensors. Error bars indicating one standard deviation are also shown as an indicator of intersubject differences. Bars filled with a horizontal pattern indicate that the mean activation is not significantly higher than a baseline level of 5%. *Significant difference between mean activations in the two different loading magnitudes.

served activity levels in biplanar conditions compared with those estimated by addition of the uniplanar levels were 11.1 ± 4.43% (mean ± SD) lower for the 50% loading magnitude and 19.3 ± 7.12% lower for the 90% loading magnitude.

Agreement in activation patterns between the 50% and 90% moment magnitude conditions was found consistently across loading conditions, with the 90% moment simply eliciting larger activation levels. On average, however, 50% MVE loading did not consistently yield a muscle response of 50% maximum, nor did 90% MVE loading consistently yield a muscle response of 90% maximum (Figure 7), although some individuals did reach these levels in a few trials for a few muscles. Furthermore, the numerical ratio required for the conclusion of direct proportionality between exertion magnitude and muscle activation was not observed. While a ratio of 1.8 is expected (90% activation ÷ 50% activation), the observed mean (SD) value was 1.31 (0.09).

In exertions where antagonist muscles could be readily identified (*e.g.*, extension and flexion), the moment magnitudes tested in this study were found to elicit higher levels of antagonist activation than reported by Lavender et al[35,36] using smaller moment magnitudes. For example, the left rectus abdominis (Figures 4a, 6a, and 8), which might be expected to be inactive at the 0° loading angle, instead exhibited near 20% activity under this condition, exceeding the 5% levels observed by Lavender et al.[35,36] In general, however, the level of antagonism observed in this experiment was not proportional to the load magnitude applied. When the moderate loading magnitudes (50%) in this experiment are compared with the high loading magnitudes (90%), obvious antagonists that are also active (>5%; *e.g.*, REO and RIO in Figures 4a and 6a) did not consistently exhibit differential responses in proportion to moment magnitude.

■ Discussion

Muscle activations obtained in this study compare favorably with those obtained by Lavender et al[35,36] under lower moment magnitudes. Although the set of muscles sampled by these authors differed slightly from the set sampled here, those muscles in the present study with a direct equivalent showed similar associations between activation and moment direction but with simply larger magnitudes (Figure 8). An increase in antagonism due to

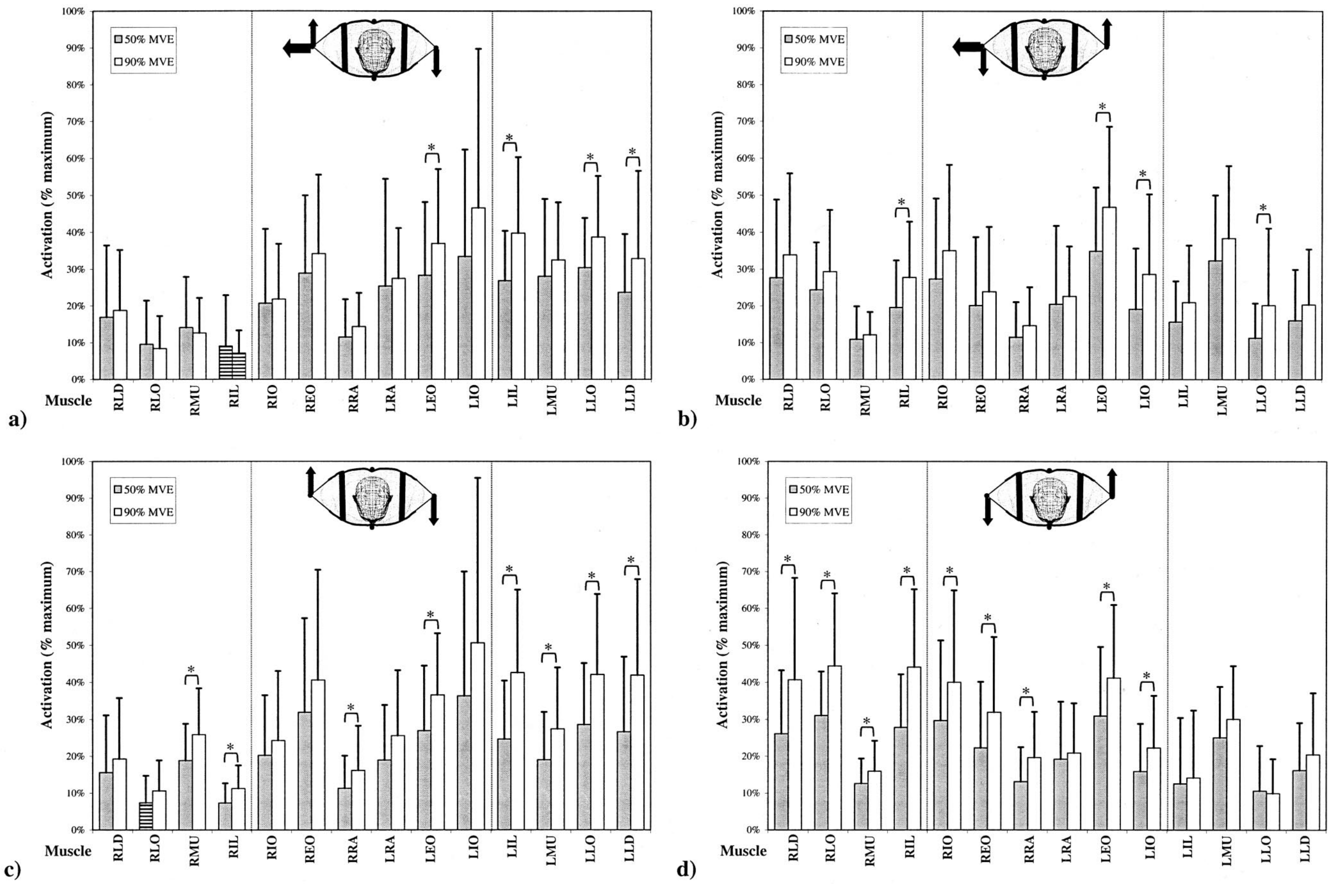

Figure 5. Normalized EMG for each muscle and moment direction under the following: **(a)** applied lateral bending with counterclockwise twist, **(b)** applied lateral bending with clockwise twist, **(c)** applied counterclockwise twist, and **(d)** applied clockwise twist. See the legend to Figure 4 for a description of symbols and labels.

increases in load seems to exist for low to moderate magnitude static conditions (*i.e.*, comparing between Lavender et al[35,36] data and the 50% loading magnitude in this study), which supports several spine stability theories.[11,18,52,60] However, this finding does not seem to apply between moderate- and high-loading magnitudes, for which little differential antagonist activity was observed. Antagonistic coactivation thus appears to reach a plateau at high-magnitude exertions, consistent with earlier findings.[18] It is possible, given an assumed association between antagonistic coactivation and trunk stability, that after a certain load magnitude the trunk has achieved a necessary level of stability and further activation of antagonist muscles is no longer necessary.

An inexact additive property between the activation levels of twisting trials and their associated nontwisting counterparts suggests that activation patterns may be adjusted based on the overall moment, rather than by a subdivision of the loads into different components with subsequent muscle activation to counteract these component moments. Although simple patterns are indeed combined to produce the more complex pattern (*e.g.*, the pattern obtained when summing activation levels for extension and twist is similar in shape to the observed pattern), there may also be an "optimization" process that reduces the net activation levels by an amount dependent on loading magnitude. Thus, although spinal loads will be higher under complex loading conditions, the levels of spinal loads would be less, by an amount dependent on loading magnitude, than the levels expected from separate consideration of the component exertions. This effect, however, can also be interpreted as support for the strong presence of nonlinearities between trunk muscle force generation and EMG output, previous reports of which have been contradicting.[46,53]

Although levels of intersubject variability in muscle activation levels were large (average coefficient of variation ~80%), intrasubject variability was substantially lower (average coefficient of variation ~30%). These large levels of intersubject variability may be the result of the use of different motor control patterns by different individuals, which in turn might play a role in the etiology of low back injury. As noted, the repetition factor was not associated with significant effects on muscle activation patterns, as determined by both ANOVA and the high ICCs (indicative of high repeatability). These results support findings obtained in occupational settings,[1] in which intertask variation in several trunk kinematics variables was much larger than intratask variation. These results also agree with a recent study[47] in

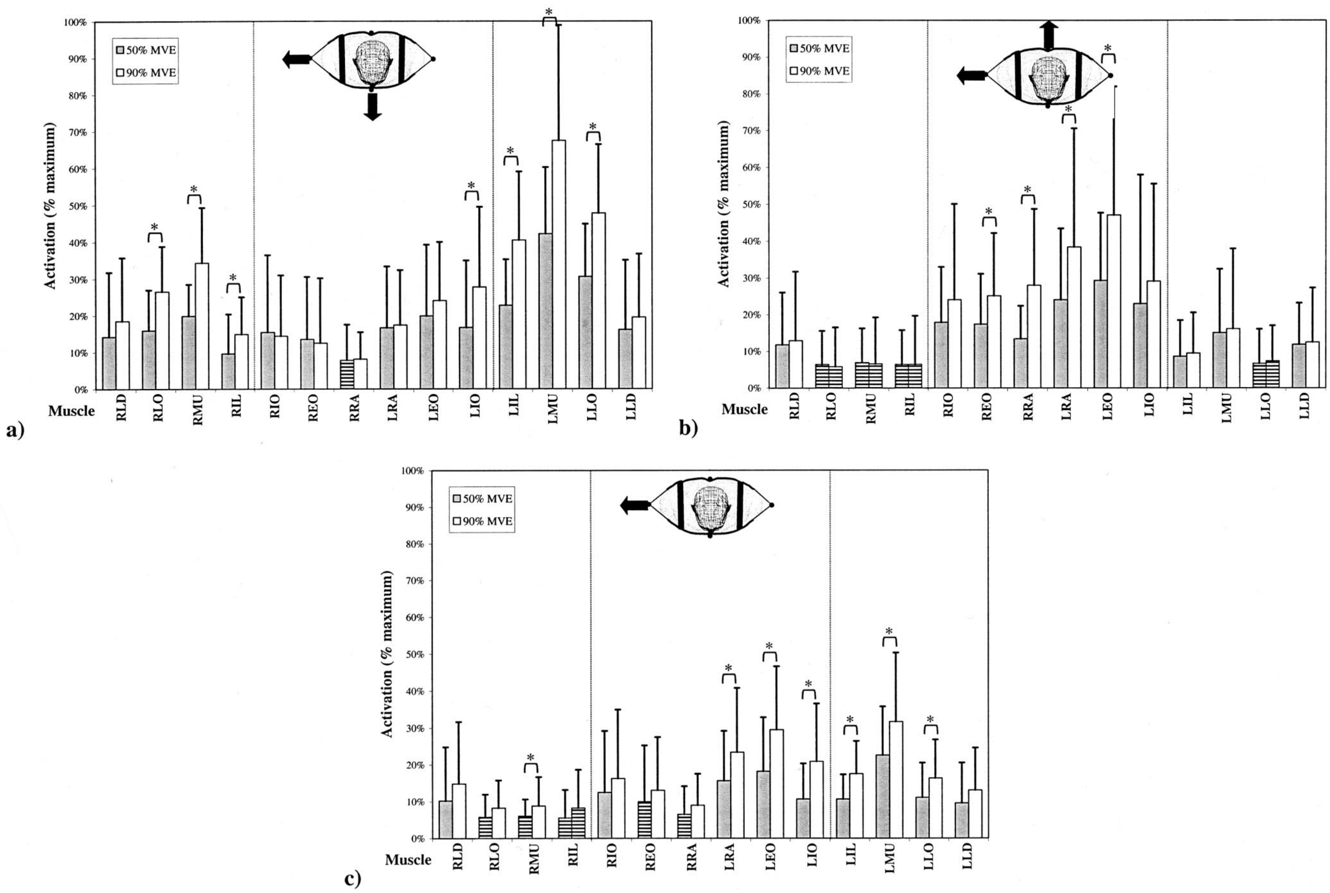

Figure 6. Normalized EMG for each muscle and moment direction under the following: **(a)** applied flexion with lateral bending, **(b)** applied extension with lateral bending, and **(c)** applied lateral bending. See the legend to Figure 4 for a description of symbols and labels.

which similar data were collected under dynamic exertions. In situations in which trial repetitions are costly, a single trial may suffice to represent muscle activity of an individual in response to specific static loads.

No significant differences in activation patterns were found between genders. Females and males were matched in terms of average heights and masses, and the required exertions were normalized as a function of each participant's capabilities. The control of participants' heights and masses was partially successful in equilibrating their MVEs (Table 1), as gender had a significant effect only for twisting MVE efforts. These controls and the observation of no differences in muscle activation between genders suggest that both genders share similar motor control strategies, at least in the limited context of static lower trunk exertions. Similar results regarding consistency between genders have been observed for torque–EMG associations during concentric and eccentric quadriceps exertions.[57] Future research may thus be able to de-emphasize gender in selecting participant groups without loss of generality, as long as independent variables related to anthropometry and strength are appropriately normalized or treated as covariates.

A notable and counterintuitive trend involved normalized average activation levels that remained lower than the normalized exertion levels. It is unlikely that this

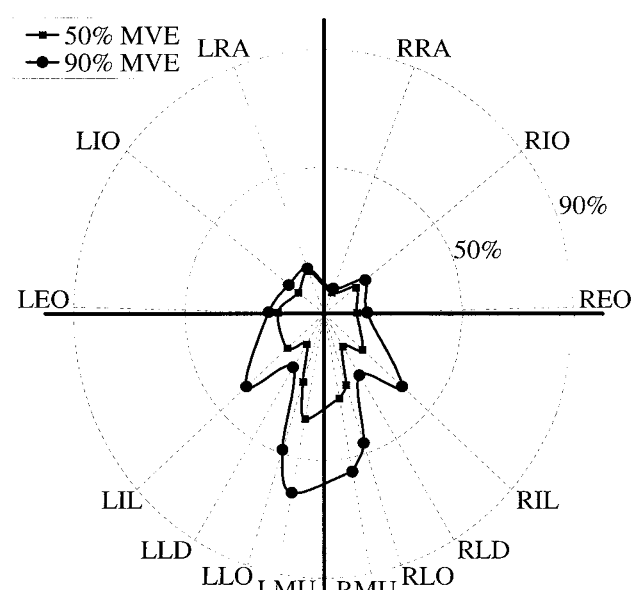

Figure 7. Polar representation of average activation patterns in applied flexion trials. The polar angles determined using a line from the L3–L4 disc center to the centroid of the muscle in a horizontal plane at that vertebral level. Circles indicate 50% activation and 90% activation, and each radial line represents one muscle. In general, and as shown for this trial, activation levels of agonist muscles were less than the expected 50% and 90% activation levels for 50% and 90% moment magnitude trials, respectively.

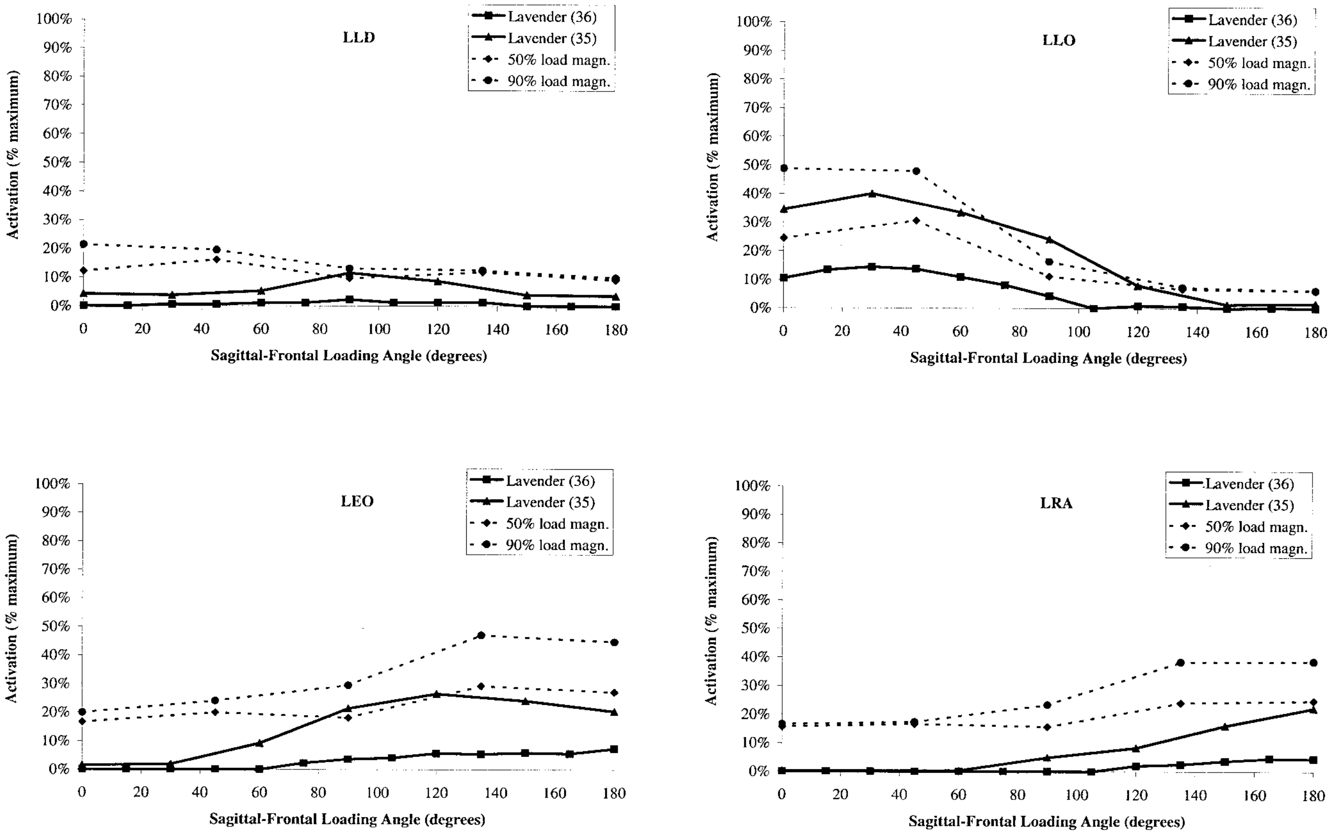

Figure 8. Comparison of selected muscles in the Lavender et al[35,36] data with the present study. Similar relationships between the two sets of data were exhibited for muscles on the right-hand side.

is a measurement error because the recorded activation levels agree well with previous studies. An alternative explanation is that the MVEs were underestimated. Average values obtained here (Table 1), however, are comparable and higher in each plane than those in Kumar et al[24,25] (overall averages: flexion 130 Nm, extension 206.5 Nm, lateral bending 118.5 Nm, twist 52.5 Nm), who used slightly older participants. The lower than expected average activation levels may have instead resulted from averaging over a range of participants, some of whom might have used different muscle activation strategies. Indeed, Nussbaum and Chaffin[48] have suggested that classification of participants based on their activation patterns can identify subgroups of individuals that use larger degrees of antagonistic coactivation and have larger levels of variability in activation patterns. Based on their results, the possibility of containing both subgroups in the participant pool of the current study is high and would explain the large intersubject variability observed. An alternative explanation is that muscles that were not monitored are active and contributing to the generation of the reactive moments. This effect could account for the very low average activation levels observed in the lateral bending exertion (Figure 6c), especially considering that the psoas muscle is recruited during lateral bending efforts.[2,7,41]

The interplay of muscles in these complex and high-magnitude exertions is difficult to quantify. However, the fact that most muscles were in an active state (NEMG $> 5\%$) across the experimental conditions suggests that synergism plays a central role in the motor control strategy used for the trunk. Whether synergism is an intrinsic part of the muscle recruitment process or a response to task or subject specific demands (*e.g.*, stability) is not yet known.

Static loads were investigated to allow for relatively precise control over lumbar moments. Static conditions further allowed for more controlled investigation of the effects of loading and individual factors because artifacts such as electrode movement, force–EMG, and velocity–EMG associations are minimal. Because dynamic loads are more relevant in applications, the focus on static loads represents the main limitation of the present study. The main findings of this study, however, are likely equally applicable to slow dynamic settings, such as heavy lifting situations, in which trunk accelerations and velocities are low and thus have little effect on applied loads and muscle force generating capabilities. This study also generated these static moments using forces applied to the upper torso. Therefore, the lumbar moments that were generated as independent variables are also combined with shear forces. Previous research on the effect of offset shear forces under similar trunk load situations, however, has found no additional activation that can be attributed to the shear loads.[8]

The current study has resulted in a new database of muscle activation patterns to complement previous efforts. Comparisons of activation patterns between and within individuals showed that intersubject variability in lumbar muscle recruitment is larger than intrasubject variability. When gender was examined, no significant differences were found between male and female patients in terms of static motor control strategies for the lower trunk. Finally, although high-magnitude exertions do require significant synergistic activation of trunk muscles, antagonistic coactivation increases from low- to moderate-magnitude exertions but remains relatively constant from moderate- to high-magnitude exertions.

Acknowledgment

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■ Key Points

- Intersubject variability in lumbar muscle recruitment is larger than intrasubject variability.
- Synergistic activation of trunk muscles is substantial under high-magnitude exertions.
- Antagonistic coactivation of trunk muscles can reach 10%–30% MVE but remains consistent under high-magnitude exertions.
- Motor control strategies for the lower trunk musculature appear comparable between genders during static trunk efforts.

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