

The Influence of Trunk Muscle Coactivity on Dynamic Spinal Loads

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Study Design. Measured trunk muscle activity was employed in a biomechanical model to determine the influence of including or neglecting muscle coactivity on predicted spinal loads.

Objectives. The purpose of this investigation was to examine the influence of muscle coactivity on spinal load.

Summary of Background Data. Electromyographic patterns in the trunk musculature have demonstrated significant levels of cocontraction during lifting exertions. Biomechanical analyses of musculoskeletal loading are often mathematically constrained from including muscle coactivity. Models that attempt to include coactive behavior are complex and difficult to implement.

Methods. Electromyographic data were collected from five trunk muscle pairs while subjects performed dynamic lifting exertions. A validated, electromyographically assisted biomechanical model was used to compute relative muscle force, lifting moment, and spinal load. Results were generated and compared from analyses that included from one to five simultaneously active muscle pairs.

Results. Trunk extensor muscles generate lifting moments as much as 47% greater than the applied lifting moment to offset flexor antagonism. Analyses that neglect muscle coactivity during dynamic lifting exertions may underestimate spinal compression by as much as 45% and shear forces by as much as 70%.

Conclusions. The level of coactive spinal loading is significantly influenced by the weight of the lifted load as well as trunk extension velocity. Muscle coactivity significantly influences the modeled load in the lumbar spine during lifting exertions and should be considered if an accurate measure of spinal loading of desired.

[Key words: biomechanics, coactivity, electromyography, spine] *Spine* 1995;20:913-919

Biomechanical models of lifting exertions have been used to analyze the loads on the lumbar spine as a function of lifting parameters. The underlying assumptions and complexity of these models vary widely, from single extensor muscle representations of trunk mechanics⁴ to multiple muscle representation using optimization criteria²⁹ and myoelectric interpretation of muscle

activity.²¹ The number of active muscles represented in a biomechanical model, and the relative contractile force in each modeled muscle is significantly influenced by the model's design and assumptions. When interpreting biomechanical analyses of spinal loads, one must determine whether or not the inclusion or neglect of modeled coactivity significantly influences the predicted results.

Muscle contractile forces act through their anthropometric moment arms to generate trunk moments while simultaneously loading the spine in compression and shear. A greater number of active muscles, that is, increased coactivity, can generate greater total force and spinal load.

Synergistic muscle activity, in concert with the prime movers, indirectly influences spinal load by changing the net mechanical advantage of the system.²² During a lifting exertion, an applied trunk moment may be developed by activating various combinations of muscles. If one assumes the body optimizes muscle recruitment, a unique solution set of muscle activity may be determined.^{1,26} Schultz and Andersson²⁹ modeled trunk mechanics by computing the muscle forces necessary to generate applied trunk moment with minimum muscle force and spinal load. The criteria may be satisfied by activating the muscles with the greatest moment arm and line of action. Sharing the trunk moment with muscles other than the optimal solution set may reduce the net moment arm, line of action, and mechanical advantage. Therefore, suboptimal coactivity patterns may require increased total muscle force, possibly resulting in increased spinal load. Documented muscle cocontraction coincident with submaximal lifting exertions³¹ indicates that the neuromuscular control system does not necessarily use only those muscles with the greatest mechanical advantage. Consequently, synergistic trunk muscle activity may influence spinal load.

Coactive trunk flexor muscles generate spinal loads through both direct and reactive mechanics. Antagonistic flexor activity during lifting exertions results in direct loading on the spine by contributing to the total force applied to the skeletal structure. The flexor activity also tends to reduce the applied lifting moment.¹⁷ To compensate, prime movers and synergists must react by generating increased contractile force and moment to prevent a decrement in the net applied trunk moment.

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Therefore, antagonistic activity generates reactive increases in spinal loads from increased extensor muscle compensation forces. In fact, spinal loading from the reactive component of antagonistic activity can be significantly greater than the direct loading because of the mechanical advantage of the trunk flexor muscles.²⁴

Existing biomechanical models of trunk loading often neglect significant muscle coactivity because of mathematical constraints. Optimization-based models of lifting have succeeded in extending biomechanical analyses generated from single muscle models⁴ to include coactive synergists.^{2,6,30} However, modeled coactivity is often inconsistent with observed levels of muscle activity during lifting exertions,¹⁶ and typically neglect antagonistic muscle activity.

Electromyographically (EMG) assisted models can simulate spinal loads from measured muscle coactivity, but they too have practical drawbacks. Spinal loads predicted from EMG-assisted models include the influence of antagonistic as well as synergistic muscle coactivity.^{9,18} Therefore, biomechanical models that attempt to account for the influences of muscle coactivity during lifting exertions^{17,21,28} predict more realistic levels of spinal loading than those which neglect coactivity. They are, however, more complex and cumbersome than many of the commonly used ergonomic models that assume only a single trunk muscle, *i.e.*, a generic "back" muscle. In addition, these models require EMG measurements, which are not always feasible to obtain in the workplace.

The complexity associated with the design and implementation of biomechanical models of lifting compels one to ask whether the influence of muscle coactivity on spinal load is statistically and physically significant. The purpose of this study was to examine the influence of muscle coactivity on modeled spinal load that occur during lifting exertions. The biomechanical literature^{16,31} demonstrates that muscle cocontraction is significant, but is its influence on spinal loading large enough to necessitate added model complexity, or is it small enough to neglect?

■ Methods

Ten male subjects with a mean height of 177.5 ± 13.4 cm and weight of 76.8 ± 8.3 kg performed isokinetic (0° , 30° , 60° , and 90° per second) and free-dynamic (fast, medium, and slow lift rates) lifts of 0, 18, and 36 kg from a trunk flexion angle of 45° to an upright posture. Electromyographic data were collected from surface electrodes, which monitored the activity of trunk muscles including the left and right pairs of the erector spinae, latissimus dorsi, internal abdominal obliques, rectus abdomini, and external abdominal obliques as reported in Marras and Mirka.¹⁶ Three-dimensional, dynamic trunk motions were measured with a lumbar motion monitor.¹³ A force plate (Bertec 4060H, Bertec Corp., Worthington, OH) and vector translation mechanics were used to measure multidimensional lifting kinetics about the lumbosacral spine.⁸

Table 1. Modeled Muscle Groups

Model	Included muscles				
	ErSp	Lat	IntO	RAbd	ExtO
1	✓				
2	✓	✓			
3	✓	✓	✓		
4	✓	✓	✓	✓	
5	✓	✓	✓	✓	✓

ErSp = erector spinae. Lat = latissimus dorsi. IntO = internal abdominal obliques. RAbd = rectus abdomini. ExtO = external abdominal obliques.

Data were input into a validated, EMG-assisted model of lifting.^{8,9} The model computes three-dimensional muscle contractile force from processed EMG-data, muscle cross-sectional area, and dynamic lines of action. Muscle forces were modulated to account for physiologic length-strength^{10,27} and force-velocity³ artifact via empirically derived relations.⁸ Spinal loads were determined from the vector sum of muscle forces. Muscle generated trunk moments were computed from the vector products of muscle forces and anthropometrically based moment arms. Model results were tested for accuracy by comparing the dynamic variability of extension moments measured from the force plate with the values predicted by the model. Model validity was examined by ensuring the predicted coefficients of force per unit muscle area necessary to satisfy conditions of dynamic equilibrium were physiologically realistic.

Each lifting exertion was modeled five separate times. First, only the EMG data from the right and left erector spinae muscles were examined. This was accomplished by zeroing the input data representing the activity in the remaining eight muscles. A second modeling run included two coactive muscle groups, the erector spinae and latissimus dorsi, with the remaining muscles set to zero. Coactivity was iteratively added to the model in this manner (Table 1) until all ten muscles, *i.e.*, five muscle pairs, were represented in the analysis.

Post-modeling analyses were performed to examine the coactive contributions to lifting moments and spinal loads. Analyses of erector spinae muscle forces and spinal loads were performed after the levels were scaled relative to the single muscle model results, then normalized relative to the maximum, *i.e.*, usually the fully coactive, value. This permitted direct comparison of results within and between muscle coactivity groups. Statistical analyses of variance were used to determine the level of significance of biomechanical factors. Modeled forces and moments from individual muscle groups as well as total applied and predicted trunk moments and spinal loads were examined as a function of lifting parameters.

■ Results

Muscle activity was found to be significantly ($P < 0.01$) greater than zero in all 10 of the measured muscles, and was statistically influenced ($P < 0.01$) by lifted load and trunk extension velocity. Post-hoc analyses (Table 2) demonstrate that activity in the flexor muscles of the trunk increased with velocity and decreased with lifted load. Extensor muscle activity increased as a function of lifted load and extension velocity.

The erector spinae muscles generated extension moments greater than the applied trunk extension moment, despite synergistic activity in the latissimus dorsi and internal oblique groups (Figure 1). Internal obliques were considered to be extensor muscles because of the lines of action defined in the EMG-assisted model,^{8,9} the electrode placement over the posterior aspect of the muscle,¹⁶ and the correlation between lifting moment and internal oblique activity (Table 2). The data indicate that the total extensor moment generated by the erector spinae, latissimus dorsi, and internal oblique groups generate lifting moments as much as 47% greater than the applied trunk moment. The excess extension moment is necessary to offset the antagonistic flexion moments generated by the rectus abdomini and external oblique muscle activity.

Force in the erector spinae muscle group required to perform a lifting exertion (Figure 2) was significantly ($P < 0.01$) influenced by the number of coactive muscles used in the model as well as the velocity of the lift (Table 3). A lifting model using only the erector spinae muscle group naturally requires that muscle group to generate the total lifting moment. A model including active erector spinae and latissimus dorsi groups shares the lifting moment between the muscles according to the relative muscle activities, sizes, and lines of action. As more coactive muscle groups are included in the model, the contractile force required in the erector spinae changes significantly. Coactive extensor muscles reduce the rel-

Table 2. Results of EMG Activity Post-Hoc Analyses

Muscle	Velocity				Force		
	0	30	60	90	0	40	80
Rt Lat	—	—	—	—	0.075	0.137	0.180
Lt Lat	0.059	0.079	0.086	0.095	0.049	0.076	0.113
Rt ErSp	0.272	0.472	0.488	0.541	0.249	0.482	0.603
Lt ErSp	0.259	0.458	0.478	0.532	0.239	0.473	0.585
Rt RAbd	0.054	0.081	0.085	0.113	—	—	—
Lt RAbd	0.016	0.048	0.053	0.085	—	—	—
Rt EObl	—	—	—	—	0.079	0.157	0.222
Lt EObl	0.092	0.160	0.164	0.197	—	—	—
Rt IObl	0.137	0.211	0.225	0.242	0.125	0.205	0.283
Lt IObl	0.062	0.106	0.113	0.119	0.049	0.111	0.140

Normalized EMG values with analysis of variance significant at $P < 0.05$ have been noted. Underlines indicate post-hoc values were not significantly different at $P < 0.05$.

Rt Lat = right latissimus dorsi. Lt Lat = left latissimus dorsi. Rt ErSp = right erector spinae. Lt ErSp = left erector spinae. Rt RAbd = right rectus abdomini. Lt RAbd = left rectus abdomini. Rt IObl = right internal abdominal oblique. Lt IObl = left internal abdominal oblique. Rt ExtO = right external abdominal oblique. Lt ExtO = left external abdominal oblique. EMG = electromyography.

Muscle Contributions to Lift Moment

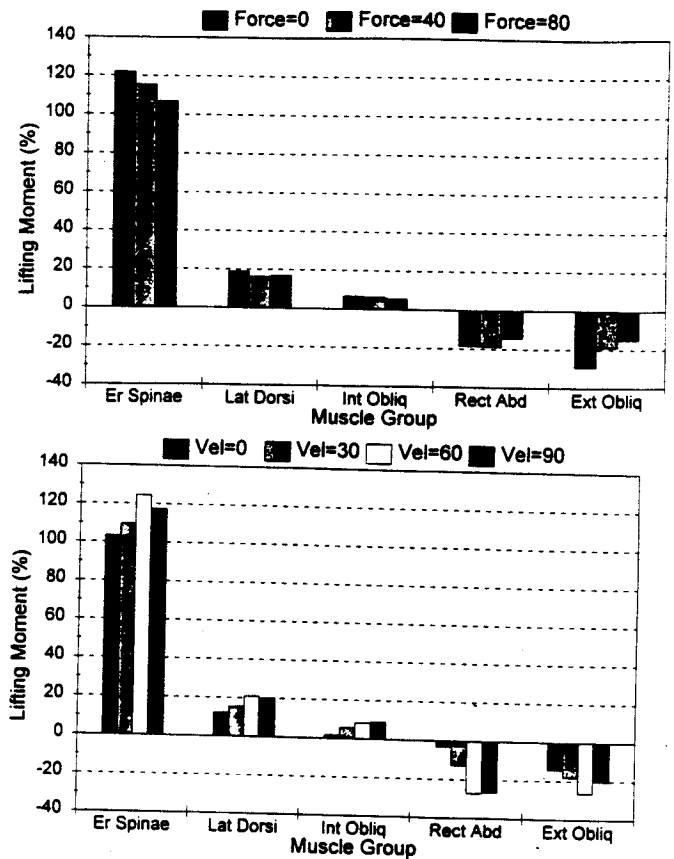


Figure 1. Lifting moments generated by each of the five muscle groups are illustrated as a percentage of the applied trunk extension moment measured from the force plate analyses. The relative extension moment generated by a muscle is a function of its activity level, size, and line of action. The percent of the total trunk moment contributed by each muscle is significantly ($P < 0.01$) influenced by the lifted load (top) and extension velocity (bottom).

ative required contractile forces whereas coactivity in the trunk flexor muscles increased required erector spinae force.

Spinal loading varied as a function of lifting parameters, as was expected from the EMG results. Predicted levels of average spinal compression ranged from 204 N during static, 0 kg exertions represented by a single extensor muscle group, to 2284 N during dynamic, 34 kg exertions represented by 10 fully coactive muscles. There was a significant ($P < 0.01$) trend for subjects to pull the weights closer to their body, reducing the necessary lifting moment as a function of trunk extension velocity. To avoid confounding the data, erector spinae force and spinal loads were described per unit lifting moment⁸ before scaling and normalizing.

Relative spinal compression (Figure 3) increases significantly ($P < 0.01$) with modeled coactivity and trunk extension velocity. Post-hoc analyses indicated that increased compression associated with the addition of synergistic muscles into the model was not significant. However, flexor coactivity greatly increased spinal com-

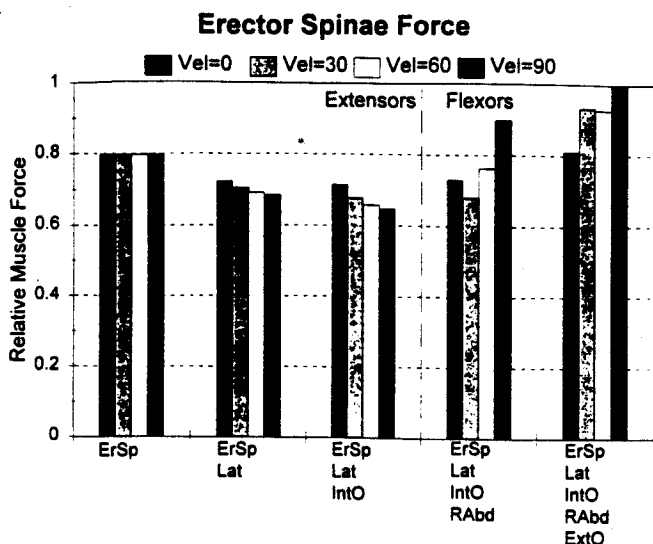


Figure 2. The force in the erector spinae muscle group is significantly ($P < 0.01$) influenced by the coactive muscles included in the model as well as the interaction between trunk extension velocity and the number of modeled muscles. The abscissa represents the muscle groups included in the model. The first set of bars represents results from a model including only the erector spinae muscle group. Additional groups are added incrementally. The last set of bars represents results from a model including all five coactive trunk muscle groups.

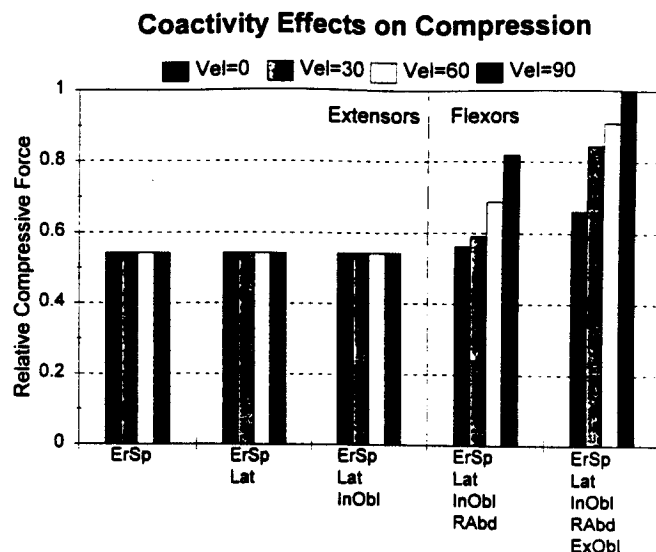


Figure 3. Compressive force on the lumbar spine is significantly ($P < 0.01$) influenced by the coactive muscles included in the model as well as the interaction between trunk extension velocity and the number of modeled muscles. A model neglecting muscle cocontraction may underestimate dynamic, coactive spinal compression as much as 45%.

pression. Dynamic lifting exertions generate compressive loads 34% greater than static loads when fully coactive modeling is employed. Models that neglect coactivity underestimate dynamic, coactive loading by as much as 45%. Anterior and lateral shear forces on the spine (Figure 4) are also influenced by muscle coactivity ($P < 0.03$). Spinal compression is also influenced ($P < 0.01$) by the interaction between trunk extension velocity and the number of coactive muscles employed in the biomechanical analysis. This indicates the relationship between spinal load and dynamic lifting may be underestimated if muscle cocontraction is neglected.

■ Discussion

There is little question that muscle coactivity exists and is significant under typical manual materials handling conditions. Analysis of EMG data collected during the lifting exertions demonstrated that all of the measured trunk muscles, including both trunk extensors and flexors, were actively contracting. In a similar experiment, Marras et al¹⁴ concluded that all trunk muscles, includ-

Table 3. ANOVA Results of Biomechanical Loads

	ANOVA			
	ErSp force (P)	Latrl shear (P)	Anter shear (P)	Compression (P)
Number of muscles	< 0.001	< 0.031	< 0.001	< 0.001
Velocity	< 0.001	< 0.945	< 0.123	< 0.001
Force	< 0.271	< 0.831	< 0.084	< 0.517
Muscle × velocity	< 0.024	< 0.852	< 0.420	< 0.001
Muscle × force	< 0.536	< 0.741	< 0.419	< 0.767
Muscle × velocity × force	< 0.886	< 0.799	< 0.687	< 0.905

Levels significantly different at $P < 0.05$ have been noted in bold. Levels significantly different at $P < 0.01$ are in italics. ANOVA = analysis of variance. ErSp = erector spinae. Latrl = lateral. Anter = anterior.

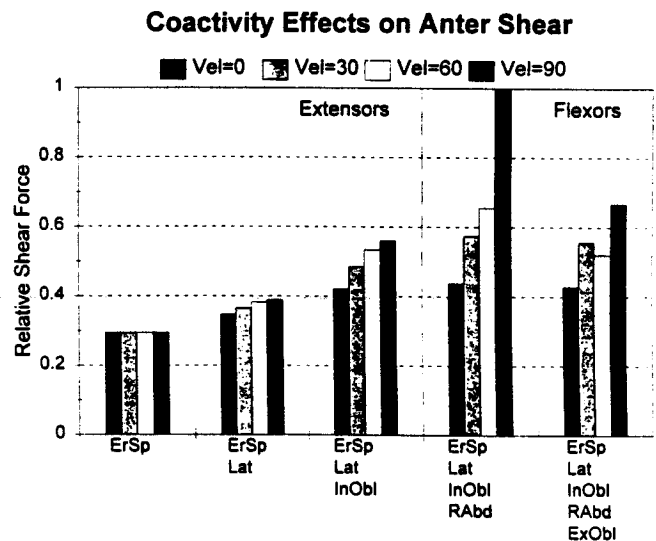


Figure 4. Anterior shear force on the lumbar spine is significantly ($P < 0.01$) influenced by the coactive muscles included in the model. A model neglecting muscle cocontraction may underestimate dynamic, coactive spinal shear force by as much as 70%.

ing trunk flexors, were active during both isometric and isokinetic extension exertions. Synergistic activity has been measured between the erector spinae and latissimus dorsi muscles as a function of maximal^{11,19} and submaximal³¹ exertions during isometric and isokinetic lifts.¹² Those and other studies^{7,14,23} have also measured antagonistic trunk muscle activity. The authors speculated that muscle coactivity may contribute to trunk stability. Trunk flexion angle, asymmetry, extension velocity, acceleration, and exertion level have been cited as significant factors affecting trunk muscle coactivity.^{15,16} Our analyses concur, indicating the relative coactive levels were influenced by the lifted load and trunk extension velocity (Table 2):

Although myoelectric activity in the trunk musculature has been measured and well documented, EMG activity does not indicate the extent to which the coactive behavior influences spinal load. Therefore, a biomechanical model was employed to examine the influence of coactivity on lumbosacral compression, shear force, and trunk extension moment. Results show that relative spinal load and applied erector spinae force increase significantly (Table 3) with trunk extension velocity. These results agree with the findings of previous studies of isokinetic¹⁸ and free-dynamic⁵ lifting exertions. The number of modeled coactive muscles also influenced spinal load and erector spinae force at $P < 0.01$.

The magnitude of the average compressive forces on the spine throughout a lifting exertion ranged from approximately 200 N during isometric, 0 kg exertions modeled without coactivity to nearly 2300 N during dynamic exertions lifting 36 kg, modeled with coactivity. Previously published compressive values represent peak levels describing lifting exertions of 10–100 kg from biomechanical models that include only the erector spinae muscles⁴ (600 N), only synergistic coactivity^{6,12,29} (450–2800 N), and those that employ a fully coactive set of muscles^{8,9,18} (2200–3200 N).

As more active muscle groups with extensor capability are included in a model, the extension moment is shared among the coactive muscles. Consequently, the contractile force in the erector spinae required to generate the applied extension moment decreases with synergistic coactivity (Figure 2). The latissimus dorsi and the posterior aspect of the internal oblique muscles share the extension moment,⁸ permitting reduced force in the erector spinae, but they also contribute to spinal compression and shear loads. Thus, reduced force in the erector spinae due to increased synergistic coactivity does not necessarily indicate reduced spinal load, for the spinal load is the vector sum of the forces in all of the active muscles. In fact, despite reduced erector spinae contractile force, post-hoc analyses indicated that the modeled compression did not change significantly with synergistic coactivity (Figure 3). Increased anteroposterior shear force with synergistic coactivity is significant ($P < 0.03$); approximately double the magnitude of

force predicted from a model that neglects cocontraction (Figure 4).

The rectus abdomini and external oblique muscle groups contribute significant flexion moments during the extension exertions. The flexion moments generated by those muscles may approach a magnitude equal to 47% of the extension moment (Figure 1). Mechanics dictate that the moment magnitude supplied by the extensor muscles must equal the magnitude of the antagonistic moment generated by the flexor muscles plus the applied trunk moment. Therefore, the total moment generated by the extensor muscles must be greater than the applied lifting moment by as much as 47%. Models that ignore the coactivity and muscle force in the trunk flexor muscles necessarily underestimate the internal moments generated by the muscles. Consequently, muscular tensile force in the extensor muscles are similarly underestimated. Inclusion of flexor muscle activity dramatically increases the modeled compression (Figure 3) and significantly increases the anterior shear force (Figure 4) on the lumbar spine. The increased spinal load is attributable to the added force of the rectus abdomini and external obliques as well as the necessary increased force in the extensors in reaction to the flexor moments.

As a greater number of coactive muscles are included in the biomechanical analysis, the relationship between lumbosacral compression and trunk extension velocity changes significantly. This interaction results from well-documented increases in trunk muscle coactivity with lifting velocity,^{15,16} specifically in the trunk flexor muscles.⁸ Thus, as trunk extension velocity increases, trunk muscle coactivity also increases, presumably to provide greater system stability.^{7,14,23} As demonstrated in this study, increased muscle cocontraction generates significant increases in spinal load. Therefore, trunk extension velocity is associated with increased spinal load, but is observable only if muscle coactivity is included in the biomechanical analyses.

This research has demonstrated that the influence of coactivity on dynamic spinal loading is both statistically and physically significant. The conclusions reached by Freivalds et al,⁵ McGill and Norman,²⁰ and Goel et al,⁶ which indicate that dynamic compressive loading can be underestimated by 30–40%, agree with the results of our analyses. Neglecting muscle coactivity may result in predicted levels of compression 45% less than dynamic coactive loads, and shear forces that are underestimated by as much as 70%.

One must be warned against adjusting spinal loads from analyses that ignore muscle coactivity by a "correction factor" to assess coactive loading. Models that ignore muscle coactivity are much more convenient to implement than those that are constrained by the measurement of trunk muscle coactivity. Clearly, it would be convenient if one could simply adjust the output of a model that does not fully account for muscle cocontraction, to predict coactive levels of spinal loading. Unfor-

tunately, the issue is complicated by the fact that muscle coactivity and subsequent spinal loads are significantly influenced by lifting dynamics, *e.g.*, extension velocity and lifted weight. Although the relationship between muscle coactivity and trunk loading may be explained via biomechanics, predicting coactive muscle force, especially antagonistic activity, has only been achieved in probabilistic terms.²⁵ Accurate predictions of the coactivity that occurs during a specific lifting exertion has not been successfully achieved in the reviewed literature.

Future research must be directed toward understanding and predicting trunk muscle coactivity during dynamic lifting exertions as a function of lifting kinetics and kinematics. Because spinal loads are strongly influenced by muscle coactivity, the ability to predict muscle coactivity would permit accurate assessment and convenient analysis of spinal loads incurred during dynamic lifting exertions.

■ Conclusions

This research demonstrated that measured trunk muscle coactivity dramatically influences the compressive and anterior shear forces on the lumbar spine during lifting exertions. Consequently, the nature of a biomechanical model of lifting, *i.e.*, the extent to which each model accounts for muscle coactivity, will significantly influence predicted spinal loads. Muscle cocontraction in the trunk flexors and extensors are significant. When this cocontraction is neglected in biomechanical analyses, there is a potential to grossly underestimate the magnitude and misinterpret the net vector direction of dynamic spinal loads.

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