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Quantifying Tissue Loads and Spine Stability While Performing Commonly Prescribed Low Back Stabilization Exercises

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Study Design. A quantitative biomechanical comparison of seven different lumbar spine "stabilization exercises."

Objectives. The purpose of this research was to quantify lumbar spine stability resulting from the muscle activation patterns measured when performing selected stabilization exercises.

Summary of Background Data. Many exercises are termed "stabilization exercises" for the low back; however, limited attempts have been made to quantify spine stability and the resultant tissue loading. Ranking resultant stability together with spinal load is very helpful for guiding clinical decision-making and therapeutic exercise design.

Methods. Eight stabilization exercises were quantified in this study. Spine kinematics, external forces, and 14 channels of torso EMG were recorded for each exercise. These data were input into a modified version of a lumbar spine model described by Cholewicki and McGill (1996) to quantify stability and L4–L5 compression.

Results. A rank order of the various exercises was produced based on stability, muscle activation levels, and lumbar compression.

Conclusions. Quantification of the calibrated muscle activation levels together with low back compression and resultant stability assists clinical decisions regarding the most appropriate exercise for specific patients and specific objectives.

Key words: lumbar spine, spine stability, tissue loading, rehabilitation, exercise. **Spine 2004;29:2319–2329**

The notion of spine stability, together with low back stabilization exercise, has become a major focus in both rehabilitation efforts and prophylactic care. The commonly used term "stabilization exercise" is a generic term that can be given to any exercise that challenges the stability of the spine while training patterns of muscle activity and spine posture to ensure "sufficient stability."[1] Sufficient stability is a concept where stability is ensured but not at overly high levels as to impose unnecessary loads on the supporting tissues.[2] Panjabi[3] detailed the preconditions for stability by describing the coordinated action of three subsystems of support structures. The first subsystem is the passive musculoskeletal subsystem where support and control of motion result from the structure of the spinal vertebral bodies and the passive stiffness of the discs, supporting ligaments and joint capsules as well as the passive properties of the muscles. Experimental evidence has shown that passive tissue damage leads to a larger "neutral zone" and joint instability. This has been observed in various low back pathologies, including disc degeneration[4] and compression fractures.[5] The second subsystem consists of the active contractile properties of the surrounding torso muscles as well as the tendons. Lastly, the neural and feedback subsystem consists of the neural control center and the mechanoreceptors located in the ligaments, muscles, and joint capsules. It is the job of the neuromuscular control system to coordinate positional and force feedback from both the active and passive structures with appropriate levels of activation to the contracting muscles in order to balance any destabilizing forces.[2,6] Hodges and Richardson[7,8] have documented that inappropriate motor patterns can result from having a history of low back troubles. Furthermore, these perturbed motor patterns have also been linked to unstable events and subsequent reinjury.[1,5] As a result, many clinicians use exercise approaches to train motor patterns for the purpose of improving spine stability. With proper technique and repetition, it is hypothesized that the subsequent motor patterns developed during the exercises will translate to more functional activities.[9] In summary, instability can both lead to back disorders but also result as a consequence of disorders and associated tissue damage.

The efficacy of different stabilization exercise approaches on reducing low back pain and dysfunction has been documented. The first and classic efficacy study assessing stabilization exercises, together with other approaches, was performed by Saal and Saal.[10] These researchers found that a nonoperative, active exercise program containing a stabilization component was successful at accomplishing a high rate of return to work (85% of the tested subjects) in subjects with a radiologically diagnosed "herniated nucleus pulposus" with radiculopathy. O'Sullivan *et al*[11] showed that a stabilization exercise program that focused on training some specific abdominal and back extensor muscles significantly reduced pain and disability levels in patients diagnosed radiologically with spondylolysis or spondylolisthesis. Our interest in the current paper is to evaluate the

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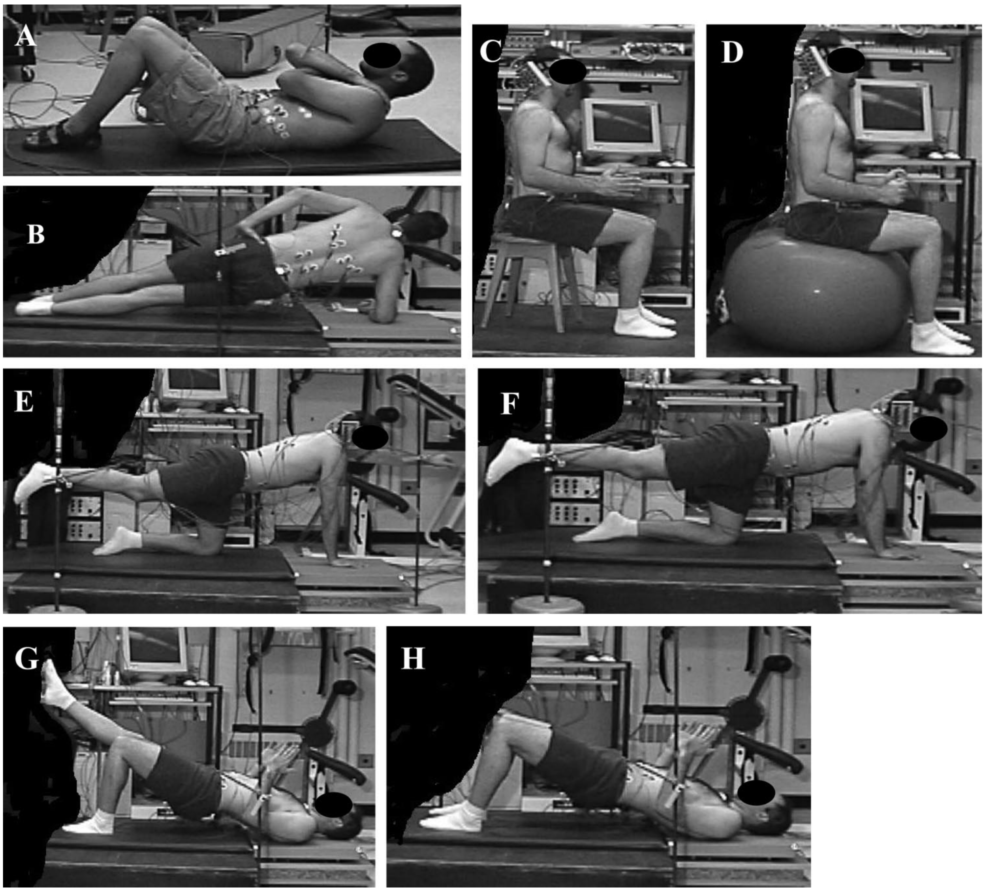

Figure 1. Pictures of different stabilization exercises: abdominal curl (**A**), right isometric side support (**B**), sitting on a chair (**C**), sitting on a ball (**D**), four-point kneeling with right leg and left arm lift (**E**), four-point kneeling with right leg lift (**F**), back bridge with right leg lift (**G**), and back bridge (**H**).

spine loading and resultant stability of some selected stabilizing exercises. Interestingly, the exercises tested in the study documented here have already been subjected to efficacy testing. Specifically, Hicks *et al*[12] reported good efficacy for success using these stabilization exercises for those patients with back pain who produced a positive result for a shear stability provocation test.

Despite the prevalence of stabilization exercises in fitness and rehabilitation programs, to date, very little research has attempted to quantify the stability in the lumbar spine resulting from the muscle activation patterns generated during specific exercises. Without a direct measure of stability available, most literature has documented muscle activation profiles, under the general assumption that as the demands on the stability of the spine increase, those muscles observed producing higher levels of muscle activity do so to provide the extra stability.[13,14] One important note is that electromyography provides insight into the actions of the neuromuscular subsystem outlined by Panjabi[3] because *in vivo* muscle recruitment strategies can be measured. However, this tool does not provide a measure of the loading characteristics of either the active or passive spinal structures, nor is stability quantified; it remains simply as an impression.

The purpose of this study was to quantify tissue loading characteristics and estimate spine stability in an attempt to provide clinicians with information to aid exercise design and prescription. The approach used in this analysis attempted to incorporate the contributions of each of three subsystems outlined by Panjabi[3] using a comprehensive biomechanical model of the *in vivo* lumbar spine.

■ Materials and Methods

Ten male subjects performed a series of eight different exercises (Figure 1) while electromyography, three-dimensional lumbar motion, and external forces were measured. These data were input into a series of biomechanical models in order to calculate a measure of L4–L5 compression and spine stability. The essential details of these methods are outlined here. The interested reader can refer to previously published papers for a detailed description. A schematic of the protocol is shown in Figure 2. All procedures were approved by the University Office for Research Ethics.

Study Participants. Ten male university students with an average age of 21 years (SD, 3 years), height of 177.8 cm (SD, 6.2 cm), and weight of 80.2 kg (SD, 12.1 kg) volunteered to participate in this study. Subjects had no history of low back pain. Before testing, subjects' age, height, weight, and breadth dimensions at the feet, ankles, knees, hips, hands, wrists, elbows, and shoulder were obtained while standing in anatomic position.

Data Collection

Exercises. Each subject performed a series of eight exercises presented in random order. The exercises (shown in Figure 1) include the abdominal curl (Figure 1A), right isometric side support (Figure 1B), sitting on a ball (Figure 1D), four-point

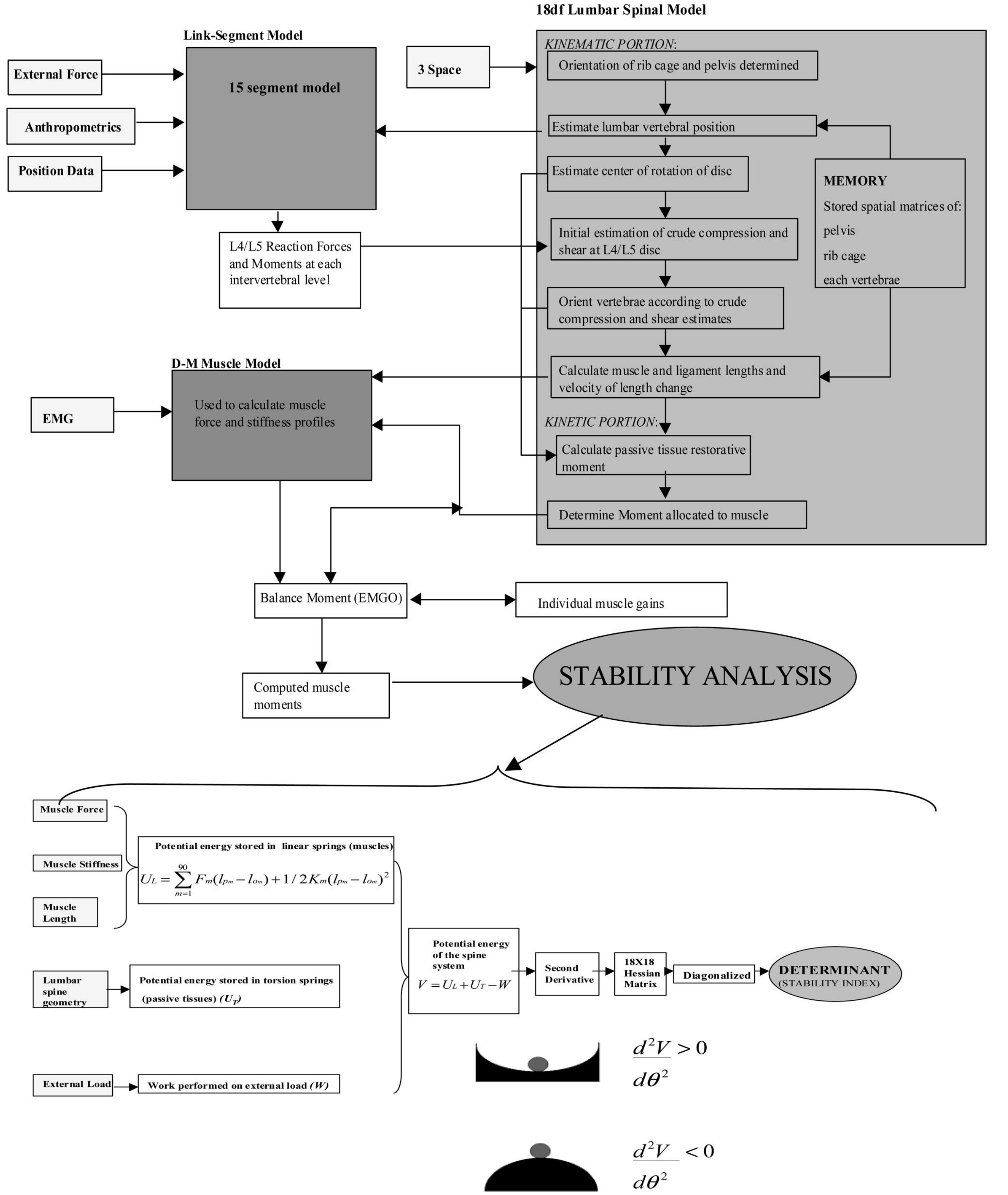

Figure 2. Flow chart of the various models used in the stability analysis.

kneeling with left arm and right leg lift (Figure 1E), four-point kneeling with right leg lift (Figure 1F), back bridge with right leg lift (Figure 1G), and back bridge (Figure 1H). To act as a control trial for the 'ball' trials and allow for assessment of unstable support surfaces, subjects performed trials sitting on a chair (Figure 1C). Each exercise was performed with a neutral lumbar spine position and controlled limb positioning. Limb and/or pelvis position was controlled through the use of an external frame with metal bars that was placed alongside body segments to act as targets. Each exercise was held isometrically for 2 seconds with an isometric contraction of the abdominal muscles (termed "abdominal brace"). A brace is an isometric contraction of all the muscles of the abdominal wall without any change in the position of the muscles. The intensity of the abdominal brace was not controlled. Study participants were given an unlimited number of practice trials, and once comfortable with the technique of performing each exercise with an abdominal brace, muscle surface EMG, three-dimensional spine posture, and external contact force measures were obtained during three successive trials.

Instrumentation

Electromyography. Fourteen channels of EMG were collected from the following muscles bilaterally: rectus abdominis, oblique internus, oblique externus, latissimus dorsi, thoracic erector spinae (longissimus thoracis and iliocostalis at T9), lumbar erector spinae (longissimus and iliocostalis at L3), and multifidus (1 cm lateral to L5). Ag-AgCl surface electrodes were positioned with an interelectrode distance of about 3 cm. The EMG signals were amplified and then A/D converted with a 12-bit, 16-channel A/D converter at 1,024 Hz. Each subject was required to perform a maximal contraction of each measured muscle for normalization of each channel. For the abdominal muscles, each subject, while in a sit up position and manually braced by a research assistant, produced a maximal isometric flexor moment followed sequentially by a right and left lateral bend moment and then a right and left twist moment; little motion took place. For the extensor muscles, a resisted maximum extension in the Biering-Sorensen position was performed. The EMG signal was normalized to these maximal contractions, full wave rectified and low-pass filtered with a second order Butterworth filter. A cut-off frequency of 2.5 Hz was used to mimic the frequency response of the torso muscles.[15]

Three-Dimensional Kinematic Positioning of the Lumbar Spine. Lumbar spine kinematics was measured about 3 orthogonal axes using a 3 Space IsoTRAK, electromagnetic tracking instrument (Polhemus Inc., Colchester, VT). This instrument consists of a single transmitter that was strapped to the pelvis over the sacrum and a receiver strapped across the ribcage, over the T12 spinous process. Thus, the position of the ribcage relative to the sacrum was measured (lumbar motion). Overall rotation of the lumbar spine was normalized relative to each subject's standing neutral spine posture. In this way, individual variance in the passive tissue contributions as a function of maximum range of motion was represented, although in this experiment there was minimal contribution of the passive tissue restorative moment because of the neutral spine posture characteristic of the stabilization exercises chosen.

External Force Measures. For exercises requiring an inverse dynamic load application, namely, the four-point kneeling exercises, back bridging exercises, and the side bridge, external force measures were recorded using an AMTI force plate. The signals were amplified to produce a peak to peak range of 20 V ($\pm$ 10 V) and then A/D converted with a 12-bit A/D converter at 1,024 Hz. Forces and moments were measured about 3 axes and were used to calculate the external force center of pressure values in the x-, y-, and z-directions. For each exercise, the subject was instructed to position the contacting segment on the force plate around the 0, 0, 0-reference point located at the center of the force plate. Reaction forces were measured at different parts of the upper body depending on the exercise being performed (Figure 1). Force plate measures were not recorded for the abdominal curl or ball sitting and chair sitting exercises. The process of using whole body linked segment dynamics and measured external forces has been explained previously.[2]

Kinematic Limb Positions. Kinematic marker data for each exercise was measured from a single subject, not part of the group of 10 mentioned above. This subject had a height of 178 cm and a weight of 79 kg. The external segment kinematics was recorded for each exercise posture with a single digital video image. The isometric position of each exercise was used to analyze the segment kinematics in the sagittal plane. The joints digitized for the kinematic analysis were the metatarsal, ankle, hip, shoulder, elbow, wrist, and hand bilaterally, as well as L4–L5 and C7–T1. The kinematic posture obtained for each exercise was controlled in the other 10 subjects with the external jig; however, the marker data were scaled to the height of each individual subject. The joint locations about the z-axis, or in the frontal plane, were scaled to the breadth measures taken from each subject guided in the space frame position jig. Since no exercise required deviations of the limbs from anatomic position in the frontal plane, breadth measures were assumed to be constant across exercises.

Data Analysis

Calculating a Stability Index. The analysis of stability was performed using a method documented by Cholewicki and McGill.[2] The calculation of lumbar spine stability is derived from the results of several interdependent models. For the interested reader, these models are described in detail by Cholewicki and McGill[2]; however, a brief description is provided here (refer to Figure 2 for a flow chart of the cascading steps involved in the stability analysis). The first model is an 8-segment link segment model that uses external force measures, subject kinematics, and anthropometrics to calculate reaction forces and moments acting at the L4–L5 intervertebral joint. The L4–L5 moments calculated with the linked-segment model are used to ultimately drive the optimization routine that determines the muscle force profiles. The reaction forces from the link segment model calculations are used in determining the shear and compression forces at the L4–L5 joint. The second model is the lumbar spine model, which consists of an anatomically detailed, three-dimensional ribcage, pelvis/sacrum, and five intervening vertebrae. More than 100 laminae of muscle and the passive tissues, represented as torsional, lumped parameter stiffness, are modeled about each axis. This model uses the measured three-dimensional spine motion data and assigns the appropriate rotation to each of the lumbar vertebral segments.[16] Muscle lengths and velocities are determined from their motions and attachment points on the dynamic skeleton of which the motion is driven from the measured lumbar kinematics obtained from the subject. As well, the orientation of the vertebral segments along with stress/strain relationships of the passive tissues was used to calculate the restorative moment created by the spinal ligaments and discs. The third model, termed the "distribution-moment model,"[17,18] is used to calculate the muscle force and stiffness profiles for each of the muscles. The model uses the normalized EMG profile of each muscle along with the calculated values of muscle length and velocity of contraction to calculate the active muscle force and any passive contribution from the parallel elastic components. When input to the spine model, these muscle forces are used to calculate a moment for each of the 18 *df* of the 6 intervertebral joints. The optimization routine assigns an individual gain value to each muscle force in order to create a moment about the intervertebral joint that matches those calculated by the link segment model to achieve mathematical validity. The objective function for the optimization routine is to match the moments with a minimal amount of change to the EMG driven force profiles. The optimization routine has a dynamic lower limit based on current activation, set on the optimized force output of the muscle to prevent any muscle from completely turning off. The adjusted muscle force and stiffness profiles are then used in the calculations of L4–L5 compression and shear, as well as spine stability. The most recent updates to the model, specifically regarding the much improved representation of the transverse abdominis, is documented by Grenier and McGill.[19] Specifically, the fascial attachment of transverse abdominis on the lumbar vertebrae was represented with 10 fascicles bilaterally on the five segments (two originating on the posterior tip of the lumbar spinous processes and the other two originating on the transverse process of the lumbar vertebrae). To capture the line of action of the fascial attachments, the 10 fascicles converge on a nodal point 60 cm directly lateral of L5 (that moves dynamically with L5).

In the final step of the stability analysis, the value for stability, or stability index, was obtained by calculating a level of potential energy in the spinal structure for each of the 18 *df* (3 rotational axes at 6 lumbar joints) resulting from the combined potential energy existing in both the active and passive spinal structures, minus any work done from external loads. The 18 values of potential energy were formed into an 18×18 Hessian matrix and diagonalized. The determinant of this matrix represented an index of spine stability. For a more detailed description of the mathematical procedures, refer to Cholewicki and McGill.[2]

Before inputting data into the link-segment model, certain modifications were made to both the data and the model to enhance the accuracy of calculations of spine load and stability for certain exercise postures. They are noted as follows:

Abdominal Curl. When performing this exercise, subjects were directed to perform a curl-up such that rotation of the upper body occurred about the base of the ribcage. Consequently, the weight supported consisted of the head and neck, thorax, and arms. Calculating moments about the L4–L5 joint would consider the entire torso mass and result in an overestimation of the flexor moment required by the muscles. To consider the true axis of rotation, the L4–L5 marker was shifted up along the long axis of the spine so to accurately represent a rotation of the thorax opposed to the trunk. A thorax distance of 0.4 m, which is characteristic of a 75th percentile male, was used. The mass proportion assigned to the thorax was 0.216 of body mass.[20] For this exercise only, the abdominal segment was considered a rigid segment and the thorax moment was then translated to the L4–L5 joint, recognizing that the rectus abdominis carries equal loading along its length.

Bridging With Single Leg Extension. In this exercise, the internal oblique does not accurately represent the activation profile of the psoas muscle because of the extended leg.[21] For this exercise, the psoas force in the lifted leg was calculated as a proportion of the moment supporting the leg, which was assumed to be primarily generated from combined action of the rectus femoris, iliacus, and psoas. The moment arms and peak isometric muscle forces used to calculate the proportions for the three listed muscles were obtained from literature[22,23] (Table 1). Then, for each subject, the support moment required to

Table 1. Parameters Used to Calculate Contribution of Psoas Muscle to the Support Moment of the Extended Leg for the Bridging Task With Leg Lift

Muscle	Peak Isometric Muscle Force (N)*	Moment Arm (cm)†	Relative Proportion of Total Hip-Flexion Moment
Psoas	370	2.9	0.19
Iliacus	430	3.0	0.23
Rectus femoris	780	4.2	0.58

*From Delp *et al.*[21]
†Moment arms are measured at the hip during the mid stance phase of gait.[28]

maintain the posture of the lifted leg was calculated. This moment was then multiplied by a proportionality constant for psoas and divided by its moment arm. The resulting force value was input into the 18 *df* lumbar spine model (Figure 2) by adding it directly to the compressive force acting on the spine, consistent with the psoas line of action.[24]

Exercise Analysis. To quantify the stability of the different exercises, the normalized EMG profiles, external forces, kinematic data, and 3-space data collected for the 2-second trial duration was input into the appropriate models. The output of interest was the total L4–L5 compression values that consider both the joint reaction forces from the link-segment model and the muscle forces from the spine model. The muscle forces and stability indexes were also obtained. For the last second of every trial, an average value of stability, compression, muscle EMG, and force was calculated. In this analysis, the optimization routine was used to balance the moments.

Several layers of sensitivity analyses were performed to assess the validity of assumptions made, together with their impact on the results. For example, to assess the differences between the muscle force profiles calculated from the measured EMG *versus* that determined by the optimization, the input data were run through the model again; however, the optimization routine was turned off. Average stability, compression, and muscle force values were calculated and compared with the results obtained with the optimized moment balance.

The second sensitivity analysis tested the assumptions made regarding the EMG-driven psoas force calculation. This analysis replaced the psoas muscle activation levels assumed to be synergistic with that of the internal obliques,[21] with activation levels reported in literature from internal measures.[25] The exercises tested in this sensitivity analysis were the abdominal curl, chair, ball, and side bridge. No measures of psoas activation exist in the literature for the other exercises performed in this study. The optimization routine was not performed in this analysis. Average stability and compression values were calculated and compared with the original method and assumptions.

While the major objective was to create a rank order of exercises based on several criteria, statistical differences were also assessed. A one-way repeated-measures ANOVA was used to identify any significant differences between the stability and compression values across the different exercises. A Tukey's pos hoc analysis assessed the significant differences ($P < 0.05$).

■ Results

Each of the 8 exercises tested produced a unique combination of moment profiles generated about the L4–L5 joint. The support moments created from both the active

Table 2. Summary of the Support Moments Created at the L4–L5 Joint to Perform the Different Exercises: Mean (SD)

	Average L4–L5 Moment (Nm)		
	Bend	Twist	Flex
Abdcurl	1.30 (1.9)	0.72 (0.99)	−56.71 (7.0)
Chair	0.54 (0.5)	0.10 (0.3)	1.47 (0.5)
Ball	0.72 (1.0)	0.18 (0.5)	1.28 (0.5)
Bridge	0.15 (3.9)	2.64 (7.6)	73.81 (32.7)
Bridge_leg	−8.42 (5.0)	−15.74 (7.6)	65.94 (33.3)
Fpn_leg	4.84 (2.9)	15.62 (8.1)	6.14 (25.3)
Fpn_arm/leg	−0.05 (5.1)	57.05 (14.6)	32.84 (23.2)
Side bridge	69.18 (21.9)	12.80 (3.9)	2.87 (3.4)

Note: In the sagittal plane, flexion is negative and extension is positive. In the frontal plane, right lateral bend is positive and left lateral bend is negative. In the transverse plane, right axial twist is negative and left axial twist is positive.

and passive structures, listed in Table 2, are required to balance the reaction moments created at the L4–L5 joint resulting from any externally applied loads. These moments are helpful in interpreting the electromyographic profiles and the loading characteristics, which are two measures that assist exercise prescription and design. Table 2 indicates that the moments experienced during both the ball and chair trials were minimal (<2 Nm) in all three planes of motion. The abdominal curl is dominated by a flexor moment (57 Nm), whereas the bridging task is dominated by an extensor moment (74 Nm). For the back bridge with leg extension, an extensor support moment of 66 Nm and a twisting moment of 16 Nm to the left, on average, were required. For the two four-point kneeling tasks, a right leg lift required an average of 6 Nm extensor moment and 16 Nm twist moment to the left. However, the contralateral arm and leg lift increased the demands on the support structures by requiring an average support moment of 33 Nm of extension and 57 Nm of left twist. The right side bridge predominately requires a lateral bend support moment (69 Nm).

Challenge to Muscles

For each of the isometric exercise postures, the average normalized EMG profiles of the abdominal and back muscles are graphed in Figures 3 and 4, respectively. The format of these graphs is intended to allow comparison of the ability for different exercises to recruit specific muscles. According to Figure 3, both the side bridge and the abdominal curl produced the highest levels of abdominal muscle activation. The side bridge produced highest abdominal activation levels, however, only on the side required to support the lateral bend moment. The abdominal curl produced high levels bilaterally. On average, the side bridge produced levels of activation of 46% maximal voluntary contraction (MVC) in the rectus abdominis, 51% MVC in the external oblique, and 57% MVC in the internal oblique, unilaterally. The abdominal curl produced about 31% MVC of activation in the rectus abdominis. In the obliques, the activation on the side of the extended leg is higher than that of the bent knee leg by about 15% MVC for the internal obliques and 6% MVC for the external obliques. On average, however, the level in the external obliques is about 27% MVC and in the internal obliques it is about 37% MVC. For these two abdominal exercises, the extensor activity is minimal compared with that observed in the other exercises.

For the extensor muscles in Figure 4, some patterns in activation occur. During the two back bridging exer-

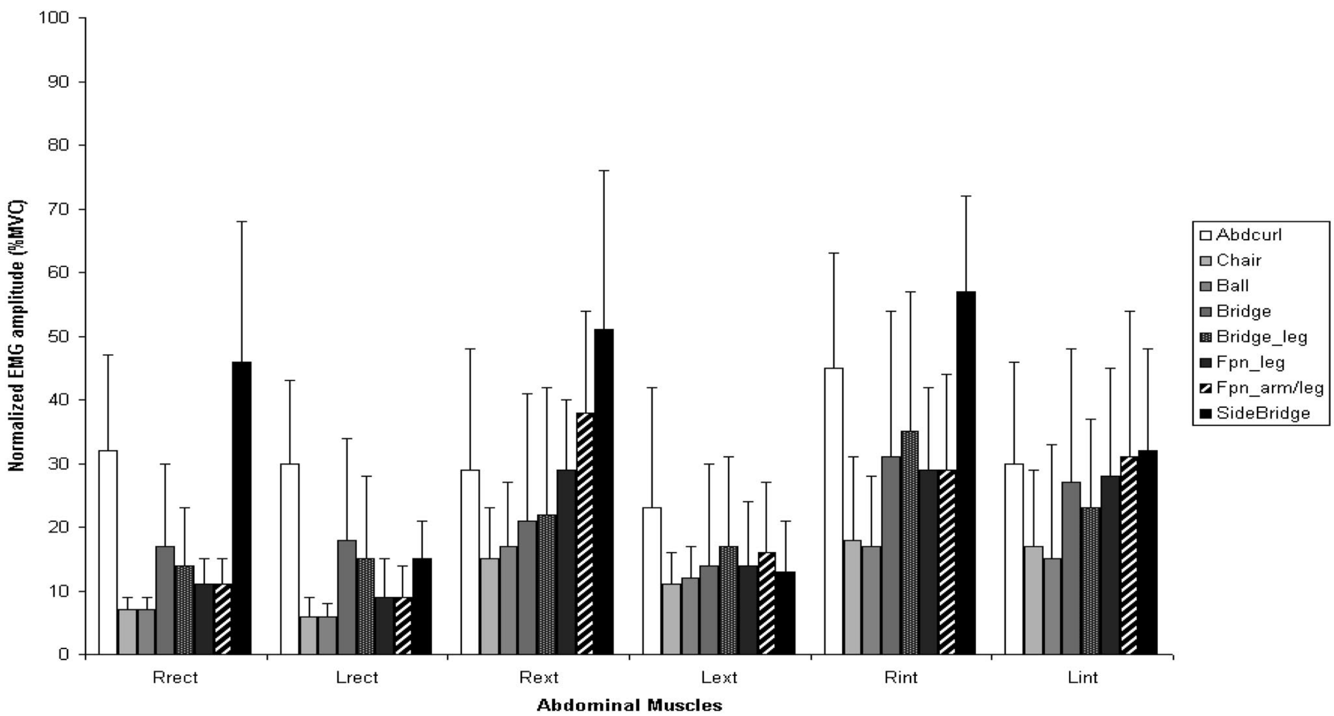

Figure 3. Average and standard deviations of the abdominal muscle EMG amplitudes for each of the exercises.

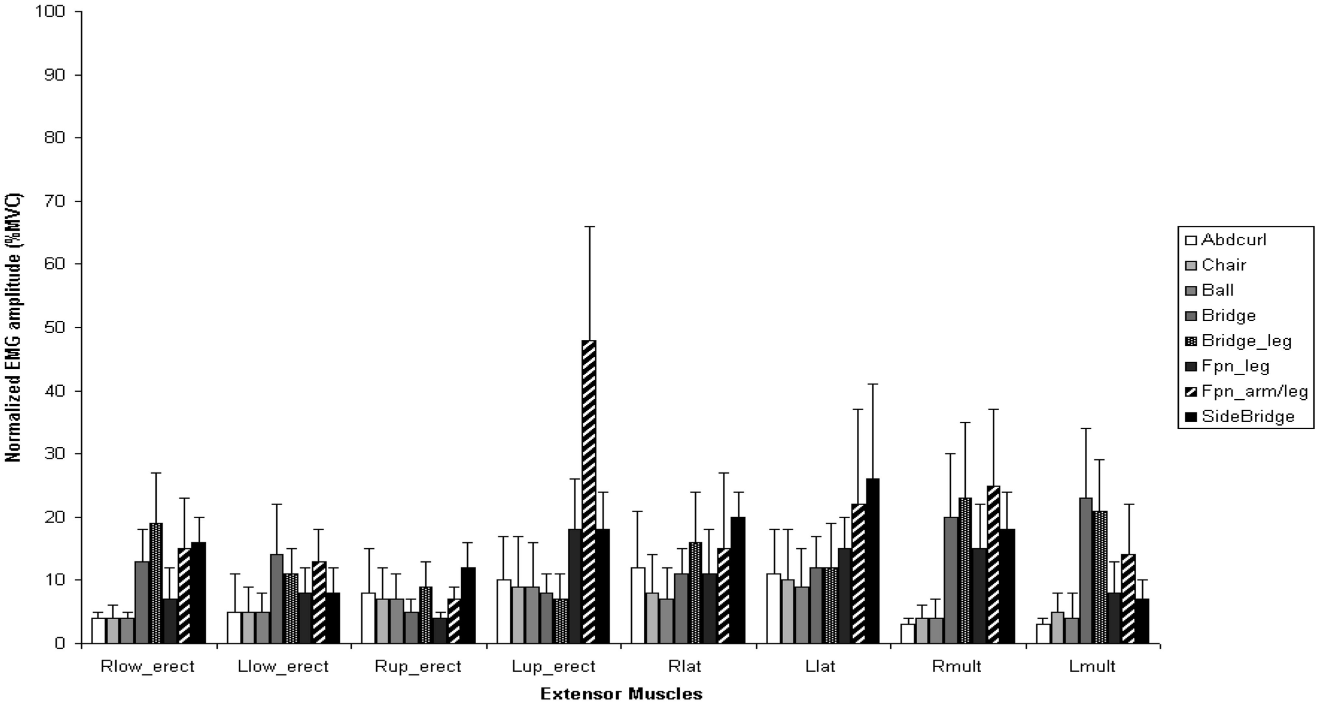

Figure 4. Average and standard deviations of the back muscle EMG amplitudes for each of the exercises.

cises, the muscle activation profiles were very similar except in the lower erector muscles. For the bridge exercise, the activation pattern was similar bilaterally at about 13% MVC for the lower erectors; however, the activation on the side of the extended leg was about 7% MVC greater than that on the contralateral side when the leg was extended. An important note is that these two exercises created the highest levels of activation in the multifidus bilaterally at about 23% MVC, compared to the other tested exercises.

During the four-point kneeling exercises, the highest levels of activation were observed unilaterally in the external obliques, upper erectors, latissimus dorsi, and multifidus. In the four-point kneeling task with right leg extension, activations were highest in the right multifidus and external oblique and the left upper erectors (15%, 29%, and 18% MVC, respectively). In the four-point kneeling task with a left arm and right leg extension, much higher activation levels were seen in these three muscles. The right multifidus was activated to 25% MVC, the right external oblique was active at 38% MVC, and the left upper erectors were active at 48%MVC.

The last two exercises are sitting on the ball and on the chair. These two exercises produced the lowest levels of activation across all muscles compared with the other exercises tested, and there were negligible differences in their activation profiles.

Challenge to Lumbar Spine Stability

Across all of the tested exercises, the ball and chair tasks produced significantly lower levels of stability ($P < 0.05$), where the level of stability for the ball was 575 Nm/rad and for the chair 582 Nm/rad. The level of spine stability was not significantly different between the two exercises. On the other end of the scale, the four-point kneeling task with contralateral arm and leg extension produces the highest level of spine stability, which was significantly greater than all the other exercises tested ($P < 0.05$). The calculated stability for this exercise was 1,386 Nm/rad. Across the remaining exercises, no significant differences exist (Figure 5 and Table 3).

Generated L4–L5 Compression

In terms of L4–L5 compression (Figure 5), the lowest level exists for the four-point kneeling task with single leg extension (2,018 N). The ball, chair, and bridge exercises produce the next lowest levels of compression: 2,097 N, 2,128 N, and 2,387 N, respectively. The fifth highest level of compression is found in the abdominal curl at 2,615 N. Although calculated differences exist between the compression values for these five exercises, they are not statistically different. The highest levels of compression are calculated for the side bridge, bridge with single leg extension, and the four-point kneeling task with contralateral arm and leg extension (2,726 N, 2,707 N, and 2,740 N, respectively). The bridge with single leg extension and four-point kneeling task with arm and leg extension produce significantly higher levels of compression than the ball, chair, and four-point kneeling with leg extension tasks. The side bridge only produces a significantly higher level of compression compared with the four-point kneeling with leg extension task.

Sensitivity Analysis

The first analysis compared the differences between the nonoptimized and optimized muscle force profiles for

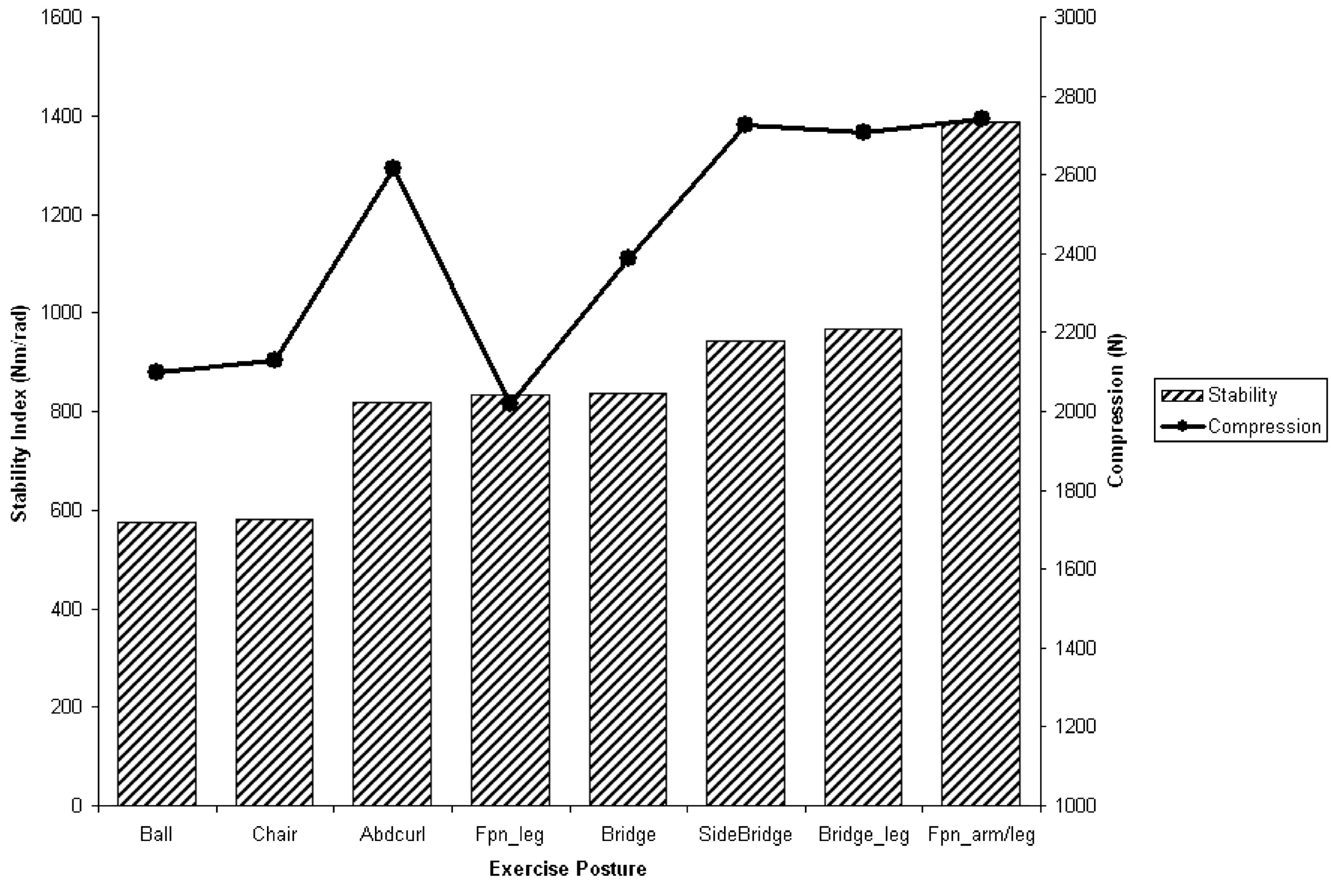

Figure 5. Stability index *versus* L4–L5 compression for each exercise posture. The exercises are ordered in terms of increasing spine stability.

two of the exercises: namely, the back bridge and the right side bridge (Figure 6). The force profiles between the two analyses were similar for most muscles; however, two main differences observed were a reduction in the levels of cocontraction in the abdominal muscles within the optimized force profiles and an increase in the psoas force in the nonoptimized force profiles. Comparing the stability and compression values between the optimized analysis and the nonoptimized analysis allows the effects of the "abdominal brace" or cocontraction pattern to be assessed (Table 4). The main observation made between the optimized and nonoptimized results is that when the abdominal cocontraction pattern was considered with the nonoptimized force profile, the values for both the stability and compression levels increased.

Table 3. Significant Differences Between Exercises for L4–L5 Compression (Gray-Shaded Area) and Spine Stability (Unshaded area).

	Ball	Chair	Abdcurl	Fpn leg	Bridge	Side bridge	Bridge leg	Fpn arm/leg
Ball							*	*
Chair							*	*
Abdcurl	*	*						
Fpn_leg	*	*				*	*	*
Bridge	*	*						
Side bridge	*	*						
Bridge_leg	*	*						
Fpn_arm/leg	*	*	*	*	*	*	*	

*Significance ($P < 0.05$).

The last analysis performed on the exercises was a sensitivity analysis to assess the effects that calculating the psoas force from the internal oblique activation had on the results of the nonoptimized analysis. According to Figure 6, the internal oblique activation levels measured during each exercise produces extremely high levels of psoas force (~ 600 N across exercises). Internal measures of psoas activation were found in previously published literature for four of the eight exercises: namely, the abdominal curl, ball, chair, and side bridge.[25] The activation levels for the sensitivity analysis and the resulting psoas force calculated by the spine model are listed in Table 5. In terms of stability, altering the psoas activation levels produced minimal changes compared with the nonoptimized results. For L4-L5 compression, however, reductions of up to 400 N were observed in both the side bridge task and the abdominal curl compared with the nonoptimized results.

■ Discussion

The analysis in this study is intended to provide clinicians with greater insight into the loads imposed on the differ-

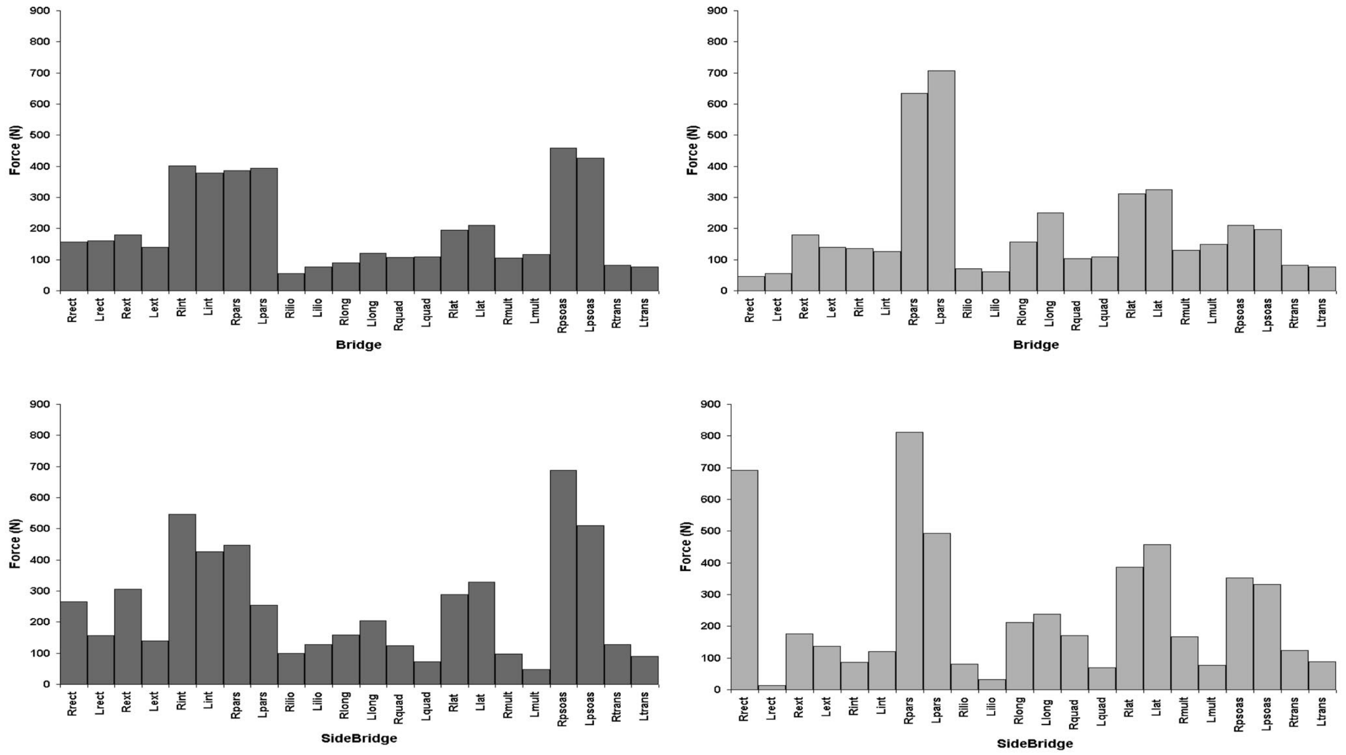

Figure 6. Nonoptimized (dark gray bar) and optimized (light gray bar) force profiles for each exercise.

ent spinal tissues and the resultant lumbar spine stability developed while performing commonly prescribed stabilization exercises. To summarize the findings of this study, a graphical representation of the continuum of these variables for the different exercises was created (Figure 7). Exercises were ranked according to the magnitude of stability *versus* compression as well as which exercises focus on training the abdominals *versus* the extensors. An important note is that goals for both preventing and rehabilitating low back pain vary across patients. Previous research has suggested that the level of endurance of the different torso muscles and the production of certain motor patterns created for spine stability are linked with lumbar spine health.[1,26] However, for some patients, the need to minimize compressive load penalties and avoid certain deviated spine postures are the predominant goals for therapy. The purpose of this continuum is to assist clinicians with making clinical decisions to better match exercises to the varying needs of the patient.

An interesting finding in this study is that there were minimal differences between the muscle activation profiles, spine stability, and L4–L5 compression levels calculated during the ball and chair tasks. At first glance, these findings appear contrary to previous literature, which has shown that activities performed on a labile surface creates higher levels of muscle cocontraction compared to the same activity performed on a more stabile surface.[27] The increased cocontraction is thought to provide the necessary stiffness to the trunk to oppose the destabilizing forces from the unstable support surface. In this study, the abdominal brace maneuver performed during the chair task appears to have been significant enough to equate the levels of stability between the two exercises and negate the effects of the unstable support surface. Previous research has indicated that an

Table 4. Summary of Average Stability and L4–L5 Compression Values for Three Exercise Analyses

	Average Stability Values (Nm/rad)			Average L4–L5 Compression Values (N)		
	Optimized	Nonoptimized	Psoas Sensitivity Analysis	Optimized	Nonoptimized	Psoas Sensitivity Analysis
Abdcurl	817 (3)	916 (3)	914 (3)	2615 (5)	3422 (5)	2998 (3)
Chair	582 (2)	714 (2)	715 (2)	2128 (3)	2853 (1)	2712 (1)
Ball	575 (1)	710 (1)	725 (1)	2097 (2)	2864 (2)	2750 (2)
Bridge	837 (5)	1031 (5)	—	2387 (4)	3231 (4)	—
Bridge_leg	968 (7)	1091 (6)	—	2707 (6)	3656 (6)	—
Fpn_leg	833 (4)	1022 (4)	—	2018 (1)	3080 (3)	—
Fpn_arm/leg	1386 (8)	1201 (7)	—	2740 (8)	3669 (7)	—
Side bridge	942 (6)	1292 (8)	1204 (4)	2726 (7)	3913 (8)	3367 (4)

Note: The rank order is listed in parentheses.

Table 5. Activation Levels and Calculated Muscle Forces for the Psoas Muscle Driven From Internal Oblique Electromyography and From Direct Internal Measures Obtained From Previous Research

Exercise	Psoas From Internal Oblique*		Directly Measured Psoas From Literature	
	Average EMG (%MVC)	Average Force (N)	Average EMG (%MVC)	Average Force (N)
Abdcurl	37	531	7	247
Chair	17	360	12	267
Ball	16	350	12	267
Side bridge	0.57(R) 0.32(L)	687(R) 509(L)	0.21(R) 0.12(L)	433(R) 310(L)

R = right; L = left.
*The internal EMG measures of psoas were obtained from Juker *et al.*[24]

abdominal brace does seem to enhance the level of stability within the lumbar spine.[19] The results of the sensitivity analyses performed this study further support this notion.

The two sensitivity analyses performed in this study were intended to assess the effects of the optimization routine on the results. Coincidentally, the information obtained from these two analyses provided insight into the effects of the abdominal brace on both spine stability and L4–L5 compression. The first sensitivity analysis showed that when the optimization was executed, lower levels of both L4–L5 compression and stability were obtained, compared with when no optimization routine was used (Table 4). This finding coincides with results from Granata and Wilson[28] who found that there is a reduction in cocontraction patterns within the force profiles, particularly in the obliques, when optimizing simply to satisfy moment requirements. However, when considering both external moments and a minimum level of spine stability in the objective functions, cocontraction patterns that more closely represented the measured EMG profiles were created in the muscle force profiles. As stated previously, the objective function for the optimization routine used in our study was based only on moment requirements. Similar to the findings of Granata and Wilson,[28] the cocontraction patterns in the abdominal muscles were reduced in the optimized force profiles when compared with the muscle activation and nonoptimized force profiles (Figure 6). Consequently, the higher levels of spine stability and L4–L5 compression calculated in the nonoptimized analysis can be attributed to the incorporation of this abdominal cocontraction.

It is very evident that not only does the abdominal brace lead to an increase in lumbar spine stability but also an increase in L4–L5 compression. Across tasks, the L4–L5 compression increased by about 1,000 N when the abdominal cocontraction pattern was considered. However, in the nonoptimized force profile, not only was the abdominal muscle c contraction pattern considered but also very large psoas forces were calculated. The second sensitivity analysis assessed the impact of this increased psoas force on L4–L5 compression and spine stability values for the nonoptimized analysis. In our methods, an assumption was made that the electromyographic profile of the internal obliques was representative of the profile for the psoas and transverse abdominis to eliminate the need for indwelling electrodes during the collection. McGill *et al*[21] reported that a range of 2% to 15% MVC error could exist within the predicted EMG levels in certain controlled tasks. The force distributions of the nonoptimized results show that the internal oblique activation profiles led to excessively high levels of psoas force across all tasks in our study. This overestimation probably occurred because subjects were asked to perform a brace during each exercise posture. The increase in internal oblique activation resulting from this maneuver may have led to a decoupling between the two

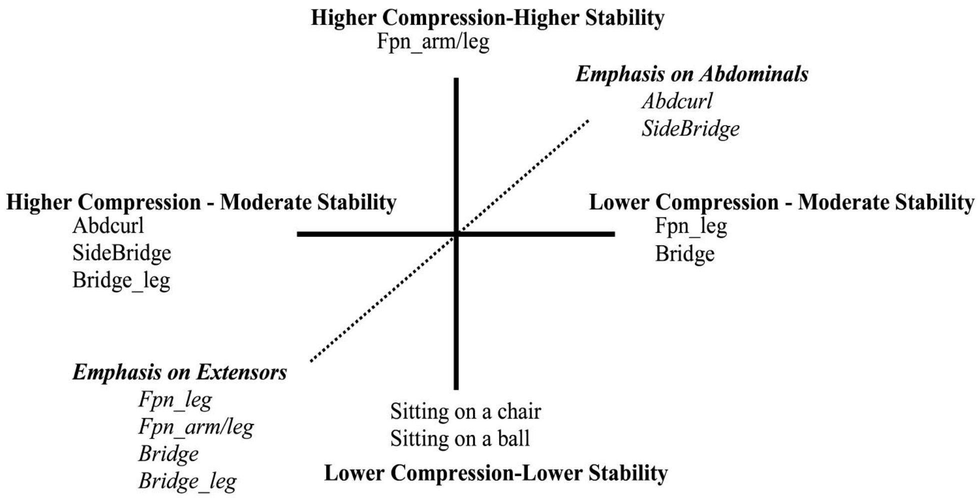

Figure 7. Exercises recommended for specific goals.

muscles not recognized before. Once internal measures for psoas activity were obtained from the literature and processed through the Distribution Moment muscle and lumbar spine models, a reduction in both the psoas forces and L4–L5 compression values were observed for each of the tested tasks; however, minimal change was seen in the stability values. After accounting for the increased psoas forces, the abdominal brace maneuver led to an increase of about 300 to 700 N depending on the exercise. However, an important note is that the intensity of abdominal cocontraction for the brace maneuver was not controlled; therefore, the increase in stability and compression that occurred as a result of the brace was highly dependent on the intensity of the contraction. The last consideration within this analysis is that only rotational stability was assessed. Translational stability involved in the controlling of shear forces was not considered in this analysis.

The role of clinicians is to assess a patient and determine the most appropriate exercise therapy, together with subsequent progression of exercise challenge over time. Knowledge of resultant muscle activity levels together with spine compression and stability will aid in the decision-making process. Without question, when prescribing exercise therapy, factors outside of those measured in this study should also be considered, including subjective reports from the patient. For example, in our clinical experience, some patients report that performing the back bridge exercise exacerbates their low back pain, but based on the data obtained here, we could not offer a specific mechanism. An important finding in this study is that an abdominal brace is a maneuver that can be performed to increase the level of spine stability during different tasks. This increase in stability is not without a modest increase in L4–L5 compression, highlighting the need to determine the loading tolerances of the patient through provocative testing.[1] It is hoped that the quantitative data documented here will assist in matching the most appropriate exercise challenge for the individual patient.

Key Points

- Using various assumptions and variations to a biomechanical model, quantification of lumbar spine stability and L4–L5 compression was performed for seven different stabilization exercises.
- Performing an abdominal muscle co contraction pattern increases lumbar spine stability and L4–L5 compression.
- Individual differences in recruitment patterns, compressive loading tolerances, and stability demands guide clinical decisions regarding exercise design and prescription.

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