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Spine Loading as a Function of Gender

William S. Marras, PhD,* Kermit G. Davis, PhD,*† and Michael Jorgensen, PhD*‡

Study Design. *In vivo* laboratory studies were conducted to investigate the spine loads imposed on men and women during a series of lifting tasks that varied in the degree of lifting control required by the subject.

Objective. To identify and understand differences in spine loading and musculoskeletal control strategies between men and women performing lifts of varying task complexity.

Summary of Background Data. Few studies have examined differences in spine loading as a function of individual factors such as subject gender. Furthermore, no biomechanical studies have attempted to quantify and understand how differences in anthropometry between genders might influence muscle recruitment and subsequent spine loads. Because the modern workplace seldom discriminates between genders in job assignments, it is important to understand how differences in spine loading and potential low back disorder risk might be associated with gender differences.

Methods. For this study, 140 subjects participated in two separate experiments requiring different degrees of musculoskeletal motion control during sagittal plane lifting. The two experiments consisted of 35 men and 35 women performing lifts in which motion was isolated to the torso and 35 men and 35 women completing whole-body free-dynamic whole body lifts. An electromyography-assisted model was used to evaluate spine loading under these conditions.

Results. Absolute spine compression generally was greater for the men. Under the highly controlled (isolated torso) conditions, most differences were attributed solely to differences in body mass. Under a whole-body free-dynamic condition, significant differences in muscle coactivations resulted in greater relative compression and anterior–posterior shear spine loading for the women.

Conclusions. Differences in spine loadings as a function of gender under the more controlled lifting conditions were primarily a function of different body masses. However, loading pattern differences existed between the genders under whole-body free-dynamic conditions as a result of kinematic compensations and increases in muscle cocontraction, with women generally experiencing greater relative loads. When spine tolerance differences are considered, one would expect that females would be at greater risk of musculoskeletal overload during lifting tasks. [Key words: electromyography, gender kinematics, lifting, low back pain, spine loads] **Spine 2002;27:2514–2520**

Recent reviews of low back disorder (LBD) causality have indicated that factors inherent to the individual, workplace, physical factors, and organization factors all can contribute to biomechanical loading and LBD risk.[53] Whereas numerous studies have evaluated biomechanical responses to workplace physical factors[10,17,19,23,24,36,43,48,56] and psychosocial factors,[11,37] few have assessed the contribution of individual factors such as gender to risk.

Work exposure has changed for women recently. Over the past several decades, women have increasingly undertaken physically demanding jobs that traditionally have been performed by men.[60] Work requirements are seldom adjusted for gender, and there is reason to expect that the biomechanical effect on the worker may differ significantly as a function of gender. For example, female lifting strength ranges between 40% and 73% of male strength.[20,29,31,33,34,52,59,62] In addition, women have lower lateral bending and axial twisting strength.[20,32,49,50,52] Whereas factors such as anthropometry may contribute to strength differences between genders, it is important that the biomechanical implications inherent to gender differences among workers be understood.

Differences in strength between the genders may influence the trunk motions, muscle activities, and subsequent spine loads. The authors hypothesize that when the primary muscles (*e.g.*, erector spinae muscles during lifting) approach their maximum capability, additional muscles (*e.g.*, internal oblique or latissimus dorsi muscles that have less mechanical advantage) are recruited to offset the external lifting moment. Although these additional muscles have the capability to offset the lifting moment, they also include a large antagonistic component because of their oblique orientation to the spine that can result in greater cocontraction increases, leading to changes in the nature of spine loading (*e.g.*, shear *vs.* compression).[21]

At this writing, few studies have investigated whether differences in spine loading exist between men and women in the performance of a physically demanding lifting task. Only 11 studies[2,8,9,13–16,30,38,56,59] were found that evaluated the spine loads for both men and women, with only four studies[8,15,16,30] reporting gender differences. These studies did not evaluate whether significantly significant differences in anatomy or biomechanical functioning exist between the genders. In two of the studies,[8,30] a simple two-dimensional static model

From the *Biodynamics Laboratory, Institute for Ergonomics, Ohio State University, Columbus, Ohio, the †Department of Environmental Health, University of Cincinnati, Ohio, and the ‡Department of Industrial and Manufacturing Engineering, Wichita State University, Wichita, Kansas.
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Address reprint requests to William S. Marras, PhD, Biodynamics Laboratory, Ohio State University, 1971 Neil Avenue, Columbus, OH 43210. E-mail: marras.1@osu.edu.
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Table 1. The Number of Subjects and Mean (Standard Deviation) for Age, Standing Height, and Body Mass for the Participants in the Three Conditions for the Study

	No. Subjects	Age (y)	Standing Height (cm)	Body Mass (kg)	Trunk Depth at Illiac Crest (cm)	Trunk Breadth at Illiac Crest (cm)	Trunk Depth at Xyphoid Process (cm)	Trunk Breadth at Xyphoid Process (cm)
Isolated torso condition								
Men	35	23.9 (4.0)	177.5 (7.4)	75.8 (12.7)	20.8 (2.4)	29.3 (3.3)	21.4 (2.7)	30.3 (2.4)
Women	35	23.1 (4.7)	166.3 (7.4)	62.2 (9.5)	19.0 (3.0)	27.2 (3.5)	19.4 (2.5)	26.9 (1.9)
Whole-body free-dynamic condition								
Men	35	22.2 (2.6)	177.8 (7.3)	80.1 (14.4)	21.6 (2.5)	30.1 (2.6)	21.9 (2.4)	31.0 (2.1)
Women	35	22.1 (3.2)	166.2 (5.8)	60.8 (9.5)	19.2 (2.1)	27.0 (2.1)	19.2 (1.7)	26.5 (1.5)

was used to estimate spine loads that considered body mass differences, but not differences in muscle actions. Two other studies[15,16] evaluated only the extensor moments generated during lifting, but not the biomechanical loading. Thus, there is a void in the body of knowledge, in that it does not offer an accurate understanding of how spine loading might differ between men and women when performing equivalent lifting tasks.

There is reason to believe that spine loading might differ between genders. Muscle anatomy differences have been reported between men and women.[44,55] One study[44] found that the cross-sectional area of major trunk-loading muscles were 31% to 39% smaller in women than in men. Smaller cross-sectional areas result in lower muscle generating capacity,[4–6] increasing the need to recruit additional muscles. Ng et al[55] noted differences between the muscle fiber directions between genders for the internal and external oblique muscles, indicating that the directions of muscle force differs as a function of gender. Additionally, Brinckmann et al,[7] reported wider pelvic width for women and men and significant gender differences in pelvic width-to-height ratios. This suggests that, proportionally, the female trunk is architecturally different from that of males. However, to date, the literature has failed to consider important differences when assessing spine loads experienced by the different genders.

Hence, the objectives of this study were twofold: to determine whether gender differences (normalized for gender anthropometry) have an impact on the spine loads for women and men, and to understand how these differences occur.

■ Methods

Approach. Two complementary experiments were designed to assess spine loading differences as a function of gender. These experiments were designed to consider two distinct issues associated with gender-related spine loading. First, are any potential spinal-loading differences a function of more than just supported moment (*e.g.*, differences in body mass)? All previous studies exploring gender have confounded anthropometry with loading. It was unclear whether differences in subject size or differences in the biomechanical architecture were responsible for any differences in spine loading. Therefore, spine loads were normalized relative to the supported external trunk moment to determine whether differences in the biomechanical systems result in loading differences when the spine is exposed to proportional loading.

Second, does the degree of control over the lifting kinematics interact biomechanically as a function of gender and influence spine loading differences? To explore the contributions of various musculoskeletal system components systematically, two separate conditions were tested: 1) lifting tasks isolating the movement to the lumbar spine, and 2) whole-body free-dynamic lifts representing more realistic lifting motions. It was believed that contrasting the two conditions would provide information about the role of body mass, gender-related differences in the musculoskeletal system, and lifting kinematics in defining spine loading differences between genders. It was hypothesized that under more controlled lifts, differences in the spine loading between the genders would arise from differences in body mass, whereas for more complex free-dynamic lifting, changes in lifting kinematics (*e.g.*, more hip motion in women) would compensate for lower trunk strength and influence muscle recruitment patterns and subsequent spine loading patterns.

Subjects. In all, 140 subjects participated in this study. Table 1 shows the number of participants in each phase of the study as well as the mean age, height, body mass, and torso dimensions. All anthropometric measures, except age, were significantly larger for the men ($P < 0.002$). All the participants were asymptomatic for low back pain.

Experimental Design. Two experimental lifting conditions were tested, each requiring different levels of movement control by the subjects. First, subjects performed sagittally symmetric lifting exertions. The exertion was isolated to the torso, with the pelvis and leg posture fixed (isolated torso condition). This isolated the lifting motion to the trunk alone and examined trunk mechanics above L5–S1. The second condition permitted movement of the pelvis and legs while the subjects performed sagittally symmetric lifts (whole-body free-dynamic condition). Other than these differences, the experimental task requirements were identical between experiments. This study permitted unrestricted movement of the whole body from the ankles up, yet maintained the sagittally symmetric lift conditions, thus permitting comparisons with the isolated torso condition. Under both conditions, the subjects began the lift in a flexed position (55° of forward flexion), then extended their torso until an upright posture was obtained (0° flexion). The subjects lifted two boxes with respective weights of 6.8 and 13.6 kg at four different isokinetic trunk velocities: 15°, 30°, 45°, and 60° per second.

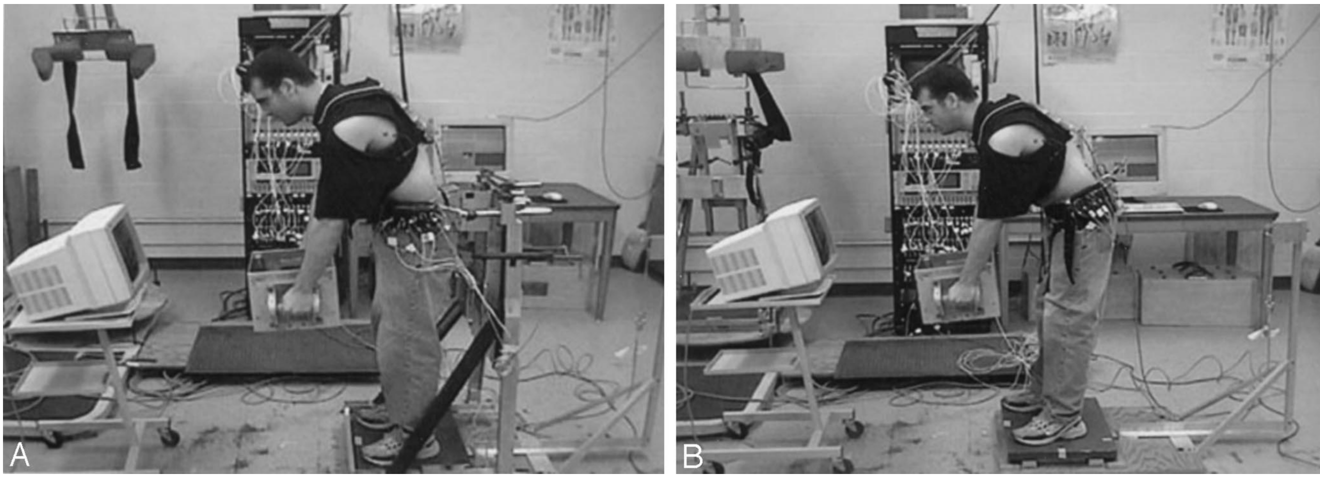

Figure 1. **A** and **B,** Subject performing the sagittal lifting tasks during the isolated torso and whole-body free-dynamic conditions.

Apparatus. The lumbar motion monitor (LMM)[39] was used to document trunk motion information and to control the sagittal velocity during the lifting conditions. The velocities were displayed on a computer monitor by providing a target in which the participants attempted to keep a trace corresponding to the instantaneous trunk position within a given tolerance (within ± 1.5% of target velocity). The subjects controlled the trunk velocity by controlling the angle of the trunk over time between 55° forward flexion to upright (0° of flexion). Under each of the conditions, the subject performed a smooth lift while keeping the trace within the displayed target. The subjects completed several practice lifts at each of the trunk velocity levels. The practice was completed after the subject completed five lifts successfully at each trunk velocity.

The muscle activities of the 10 trunk muscles (right and left muscle pairs of latissimus dorsi, erector spinae, rectus abdominus, external obliques, and internal obliques) were collected through bipolar electrodes using standard electromyographic (EMG) techniques.[54] Standard locations of electrode placement were used.[45] The EMG signals were preamplified, high-pass filtered at 30 Hz, low-pass filtered at 1000 Hz, rectified, and processed *via* a 20-ms sliding window hardware filter. Postprocessed data were collected at 100 Hz.

A force plate was used to monitor kinetic performance during lifting. During the isolated torso condition, a pelvis support structure (PSS) was used to lock the pelvis of the subject into a constant position and restricted motion during the lift (Figure 1A). Because the position of L5–S1 relative to the center of the force plate was constant throughout the entire lift, the moments and forces measured at the force plate could be accurately rotated and translated to L5–S1.[25] During the whole-body free-dynamic condition, a series of electrogoniometers were used to measure the position of the L5–S1 accurately relative to the center of the force plate as well as the participant's pelvic–hip orientation corresponding to hip tilt and rotation (abduction) (Figure 1B). The forces and moments were translated and rotated from the force plate to L5–S1 by established methods.[18]

Spine Loading Assessments. The spine loads estimated during the various experimental exertions were computed using an EMG-assisted biomechanical model developed over the past 17 years. These calculations and have been reported in detail previously.[12,22,23,40–42,46–48] The EMG-assisted model has been customized recently so that it more accurately represents gender-specific anthropometry. Fourteen gender-specific trunk muscle cross-sectional areas assessed at 11 spine levels and 14 gender specific muscle lines of action were derived from magnetic resonance imaging (MRI) while fiber orientations were taken into account.[28,44] Model incorporation of estimates for cross-sectional areas, muscle origins, and insertions of the 10 trunk muscles for men and women separately provides the most realistic representation of the trunk muscle mechanics so far, and also accounts for anatomic differences between genders. Using these new anatomic estimates, the length–strength and force–velocity relations were developed for men and women using techniques previously reported in literature.[12,23,57] Thus, the current alterations to the EMG-assisted model improved on the previously published EMG-assisted model by acquiring more accurate representations of the muscle anatomy.

Model performance was assessed through three measures: gain, R^2, and average absolute error (AAE). For each lifting trial, muscle gain was calculated by comparing the measured moments and the predicted moments while satisfying dynamic equilibrium. To be physiologically valid, muscle gains should range between 30 and 100 N/cm.[2,51,58,61] The R^2 measures the ability of the model to predict trunk moments over time, with a high value indicating that the model accounts for the variability in lifting moment, whereas AAE provides a measure of the magnitude error between the predicted and measured trunk moments. The predicted muscle gains were nearly identical for each of the lifting conditions, with values for men and women approximating 35 and 45 N/cm^2, respectively. High R^2 and low AAE error values further indicated high model fidelity. The average R^2 values were above 0.91 for the men and 0.85 for the women, whereas the AAE values were below 12 Nm for both genders in both conditions.

Data Analyses. Descriptive statistics were computed, providing means and standard deviations as a function of gender and the various combinations of independent variables. Repeated measures analysis of variance (ANOVA) statistical analyses were performed on all the dependent variables. For all significant independent variables, *post hoc* analyses (Tukey multiple

Table 2. Summary of Significant Analysis of Variance Findings for Two Conditions of the Study

		Normalized Loading		
Independent Variables	Sagittal Trunk Moment	Compression Force*	Lateral Shear Force*	A–P Shear Force*
Isolated torso condition				
Gender	0.0001†	0.33	0.55	0.01†
Gender*Velocity	0.002†	0.22	0.05†	0.46
Gender*Weight	0.0001†	0.15	0.58	0.22
Whole-body free-dynamic condition				
Gender	0.002†	0.50	0.24	0.03†
Gender*Velocity	0.98	0.0005†	0.01†	0.06
Gender*Weight	0.02†	0.19	0.82	0.80

* Spine forces normalized to sagittal trunk moment.
† Statistically significant effect ($P < 0.05$).

pairwise comparisons) were performed to determine the source of the significant effect or effects ($\alpha < 0.05$).

Results

Statistically significant differences ($P < 0.007$) in absolute spine compression between men and women were noted under both the experimental conditions, with men experiencing greater compression. Overall, the mean compression for men was about 2700 N, as compared with approximately 2300 N for women. However, once the spine loadings were normalized relative to external moment, no compression differences were present in the isolated torso experiment. The differences were largely proportional to the differences in sagittal trunk moments between men and women, and thus a function difference in body mass.

A summary of the statistically significant differences for the normalized spine loading and sagittal trunk moment as a function of gender and the gender interactions is shown in Table 2 for both experimental studies. As expected, under all test conditions, the sagittal trunk moments experienced by men were significantly greater than those experienced by women (about 18% for both conditions). The mean "peak" sagittal trunk moments were approximately 97 ± 17.6 Nm and 82 ± 24.3 Nm for women and 115 ± 23.2 Nm and 97 ± 25.7 Nm for men for the isolated torso and whole-body free-dynamic conditions, respectively.

The normalized spine loading measures in Table 2 also identified differences in spine loading between genders unrelated to differences in body mass. Few significant differences as a function of gender alone were noted except for anteroposterior shear, which was greater for women (∽11%). However, statistically significant differences were noted as a function of the interactions with gender. Figure 2 indicates that normalized compressive forces were greater for women up to trunk motions of 45° per second under the whole-body free-dynamic conditions, whereas, men experienced greater normalized compression at velocities, exceeding 45° per second. Women had about 11% higher normalized lateral shear force during the slower velocity lifts (15°, 30°, and 45° per second) than men under the isolated torso condition. These gender-related differences in normalized lateral shear increased to approximately 20% for the same lifting tasks under the whole-body free-dynamic conditions.

Hence, these results indicate that once differences in body size are accounted for, differences in spine loading between genders still remain. Furthermore, these differences are uniquely associated with differences in the degree of control required and the specific task requirements.

Discussion

It is evident from the results that biomechanically, women are not simply proportionally scaled down versions of men. Although men exhibited greater absolute compressive loads during both of the experiments, the magnitude of the differences depended greatly on the magnitude of the external moment supported by the subject, reflecting differences in body mass between genders. However, the degree of control required by the body also influences biomechanical response as a function of gender.

Under the condition requiring the least amount of subject control (isolated torso), men experienced 16% more absolute compressive loads than women, and this difference was directly related to the sagittal trunk moment supported. Because the lower body was restricted during this condition, upper body mass strongly influenced the resulting spine loads. When the loads were normalized to either body weight or trunk moment, the statistical differences between the genders disappeared. Hence, if differences in normalized loadings are present between genders, they are not related to differences in biomechanical functioning of the musculoskeletal system when exposed to the same relative kinematic conditions.

The less restrictive testing conditions (whole-body free-dynamic) indicated that as more kinematic freedom was permitted, significant differences in loadings between genders were evident and the nature of the relation between spine loads and trunk moment became more

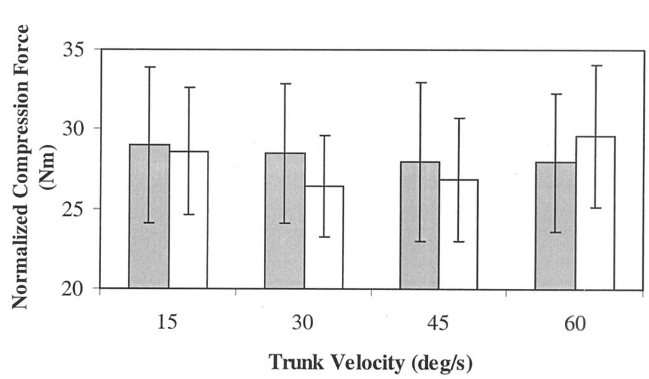

Figure 2. Normalized compression force relative to sagittal moment for men and women as a function of trunk velocity during the sagittally symmetric free-dynamic condition.

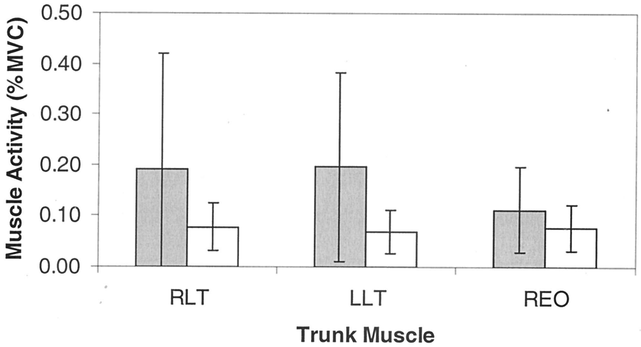

Figure 3. Normalized muscle activity for men and women for the left latissimus dorsi (LLT), right latissimus dorsi (RLT), and right external oblique (REO).

complex. During the whole-body free-dynamic condition, the normalized spine compression was greater for women during the slower conditions (*e.g.*, 30° and 45° per second), whereas men experienced greater higher normalized loads during the fast lifting condition (*e.g.*, 60° per second). Furthermore, differences remained when the compressive loads were normalized to body mass during this condition ($P < 0.02$), indicating that the resulting loads were a consequence of factors other than body size differences.

Hence, once differences in body mass were considered (normalization), no differences in spine loads between the genders were present when exertions were confined to the torso. However, when the lifting tasks involved whole-body free-dynamic kinematics, loading differences between the genders occurred as a result of kinematic compensations. Women flexed their hips about 6° more ($P < 0.01$) and had about 8° per second more hip motion ($P < 0.002$) during the whole-body free-dynamic conditions than the men. The greater reliance on the pelvis among the women may reflect the limited strength capacity in the lumbar region that has been reported previously in literature.[20,29,31,33,34,52,59,62] In the current study, women were found to have 30% less extension strength than men during the maximum exertions preformed for normalization of the EMG signals.

Muscle coactivity patterns also played a role in the spine loading differences between men and women. Women exhibited significantly greater activity ($P < 0.01$) in the latissimus dorsi (∽12% of MVC) and right external oblique (∽3% of MVC) muscles than men during the whole-body free-dynamic conditions (Figure 3). This indicated that women recruited muscles other than the primary agonist muscles (erector spinae) to complete the lifts. The recruitment of additional secondary agonist muscles, such as the latissimus dorsi muscles, would increase the coactivity, resulting in spine loads similar to the spine loads of the men because these muscles are less able to offset the sagittal trunk moments. The level and nature of the spine loads also would be affected by differences in the muscle anatomy[28,44,55] between genders (*e.g.*, muscle area, line of actions, and moment arms of the muscles). Recruiting more obliquely oriented muscles in addition to the erector spinae muscles results in more complex loads involving shear and compression. Because external oblique muscles are antagonists, they serve as trunk stabilizers, and do not actually contribute to extensor moment generation. This also increases the coactivity, which may have resulted from the limited strength capability of the women, causing the trunk to activate additional muscles to increase stability.

Occupational Risk

Although women experience lower absolute spine loads than men during both conditions, women also have lower spine tolerances,[26,27] and thus may be at greater risk of a low back injury. A comparison of spine loads relative to the tolerance limits was performed to obtain a more realistic estimation of the risk of LBD as a function of gender.[27] When the loads were expressed as a percentage of the compression tolerance, a different picture of the loading variables was revealed. Women had compression forces closer to their expected tolerances. The compression loads of the women were approximately 47% of their tolerance, as compared with men, whose compression values represented about 38% of the tolerance value. On the basis of these tolerance values, women would be expected to be at a substantially higher level of risk than men when performing identical lifting tasks. Epidemiologic research supports this finding inasmuch as women were found to be injured more often than men when performing similar heavy physical jobs.[1,3,35]

Study Considerations

Several potential differences between this study and actual working conditions should be considered in the evaluation of these results. First, tolerance limits were predicted from an existing study based on cadaver values.[27] These values provide relative benchmarks rather than absolute values. Furthermore, the regression equations relate tolerance values to the individual's gender and age, which may neglect the impact of other individual factors (*e.g.*, diet, genetics, physical exercise). Thus, these tolerance values provide only a rudimentary estimation of the spine's tolerance.

Second, subjects were required to perform controlled trunk motions during both experimental conditions. Thus some realism may have been lost. However, the degree of kinematic control was necessary to allow for more direct comparisons between the genders to determine whether inherent differences exist that may result in different spine loads. More real-world conditions may complicate the resulting spine loading.

■ Conclusion

The current study provides a comprehensive biomechanical evaluation of the differences between men and women. Men exhibited larger absolute spine loads than women. However, once differences in body mass were considered (normalization), differences in spine loading between genders were primarily a result of increased demand for subject control over the kinematics of movement. Under these normalized conditions, women generally experienced greater relative spine compression loads. They adopted a lifting style that used more hip–pelvic motion, whereas men performed the lifts using more trunk motion. The results indicated that differences in spine loading between genders is a function of the anatomic differences in trunk muscle sizes and lines of actions of the muscles as well as a combination of trunk kinematics and coactivity patterns. Women also are believed to be at more at risk relative to their loading tolerance values.

■ Key Points

- Men experience greater absolute spine loading because of greater body mass, whereas women experience greater relative loading because of more kinematic compensations.
- When spine tolerances are considered, women are more at risk of injury.

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