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A Biomechanical Assessment and Model of Axial Twisting in the Thoracolumbar Spine

W. S. Marras, PhD, and K. P. Granata, PhD

Study Design. Measured trunk kinematics, applied moments, and trunk muscle activities were employed in a biomechanical model to determine load experiences by the spine during dynamic torsional exertions.

Objectives. The purpose of this investigation was to examine the influence of dynamic twisting parameters on spinal load.

Summary of Background Data. Axial twisting of the torso has been identified as a significant risk factor for occupationally related low back disorders. However, previous studies have had difficulty describing how twisting is accomplished biomechanically, or how the spine is loaded during twisting motions.

Methods. Electromyograph activity of 10 trunk muscles was monitored while 12 subjects performed twisting exertions under various conditions of force, velocity, position, and direction. An electromyograph-assisted biomechanical model was developed to interpret the effects of those twisting parameters on spine loading.

Results. Significant flexion-extension and lateral moments were generated during the twisting exertions. Muscle coactivity associated with twisting exertions was significantly greater than that associated with lifting exertions. Employing electromyograph data to represent muscle coactivity, the model accurately predicted trunk moments and hence was assumed to reasonably reflect spine loading.

Conclusions. Under the conditions tested, the results indicated that relative spinal compression during dynamic twisting exertions was twice that of static exertions. Spine loading also varied as a function of whether the trunk was twisted to the left or right and according to the direction of applied torsion—i.e., clockwise or counterclockwise. The results may help explain, biomechanically, why epidemiologic findings have repeatedly identified twisting as a risk factor for low back disorder. [Key words: biomechanics, lifting, spinal load, twist] Spine 1995;20:1440–1451

Axial twisting of the torso has been identified by several epidemiologic studies as a significant risk factor for occupationally related low back disorders (LBDs).[1,15,40,21]

From the Biodynamics Laboratory, The Ohio State University, Columbus, Ohio.
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The U.S. Department of Labor[46] reported that twisting and turning was associated with an LBD event in 33% of workers. Snook[45] reported that 18% of worker's compensation costs were associated with twisting activities.

Although twisting of the torso has been identified as a significant risk factor for LBD, few biomechanical studies have attempted to understand how twisting is accomplished biomechanically or how the spine is loaded during twisting motions. Studies of the electromyograph (EMG) activity of the trunk muscles have described significant muscle coactivation.[2,4,50] Carlsoo[4] noted that many of the coactivating muscles were not oriented so that they contributed to twisting torque. Pope et al[38] reported the largest amount of EMG activity occurred in the agonist internal and external oblique muscles during twisting. However, they also noted a high degree of coactivation of the antagonist muscles as well as the erector spinae and rectus abdominis muscles. In a separate publication, Pope et al[39] found the bilateral symmetry of the internal oblique and rectus abdominis muscles changed significantly when the trunk was pre-rotated to either side. Furthermore, the maximum torque increased when the trunk was pre-rotated away from the direction of the twisting effort. McGill[25] observed the EMG activity in six trunk muscles while subjects performed isometric and isokinetic (30 and 60 deg/sec) torsional exertions. Significant latissimus dorsi activities were noted in the study, and myoelectric activity was lower in the isokinetic trials than in the isometric exertions.

Biomechanical models could help facilitate the understanding of spine loading and determine whether the role of muscle activity can be accounted for during trunk actions. Pope et al[38] and McGill[25] have attempted to model trunk musculoskeletal activity during torsional exertions. The models used electromyography to determine the level of force in each of the trunk muscles, but were able to account for the twisting moment only by permitting the muscle force per unit area to far exceed realistic levels. The models required muscle gains of 220 to 280 N/cm^2, whereas physiologic measurements[30,41,48] ranged from 30 to 100 N/cm^2. Hence, it was unclear whether they had identified all of the trunk muscles that significantly contribute to torsional moment. Another possibility is that previous models of trunk torsion misrepresented the size of muscles or lines of action, thereby underestimating their force capacity. This possibility is supported by the conclusions of McGill,[25] who noted that reasonable gain values (i.e., 84 N/cm^2) were achieved when the force and activity in several muscle groups were overestimated by normalizing relative to submaximal activity in those muscles. Because these models were not able to accurately assess muscle forces, they also were incapable of accurately estimating the loads experienced by the spine during axial twisting.

The Pope et al and McGill studies controlled or measured axial torque only. However, Parnianpour et al[36] reported large coupled torques. Their subjects produced 63% of their maximum sagittal plane strength and 44% of their frontal plane strength while maximum axial torque was applied. Hence, significant moments are generated in all three dimensions during torsional exertions. Measured, three-dimensional trunk moments and forces are necessary for interpreting the role of muscle activities during the exertions and can provide insight into the biomechanics and epidemiology of twisting tasks.

The objectives of the present study were to document kinetic coupling and myoelectric activity and to understand and model the exertion to develop realistic simulations of spinal loads during torsional exertions. Specifically, we wanted to measure and separate pure torsional moments from confounding accessory trunk motions during twisting exertions and then document relative muscle activity to develop an understanding of the behavior of trunk muscles during isometric and isokinetic torsional exertions. As the final step, we would model the activity of the trunk musculature to gain an appreciation for how muscle activities contribute to spine loading during twisting exertions.

Methods

The Experiment. To explore the activity of the trunk musculature during torsional exertions and axial twisting, an experiment was performed that attempted to control the posture, force, and motion characteristics of the trunk. The experimental trunk position and trunk motion characteristics were derived from an industrial database and represent trunk motion characteristics commonly seen in industry.[21] While subjects were performing these exertions, EMG activity of the trunk musculature was recorded and used to determine trunk loading via an EMG-assisted model.

Subjects. Twelve men, 21 to 31 years old, participated in the experiment. None had a history of low back disorder and each participated in a training session on a date before experimental testing to become familiar with the experimental protocol. Gross anthropometric characteristics were collected for all subjects. Mean weight (± standard deviation) of the subjects was 76.4 ± 8.4 kg, and mean height was 177.0 ± 16.4 cm.

Apparatus. The experimental apparatus used to test the subjects and control the experimental conditions is shown in

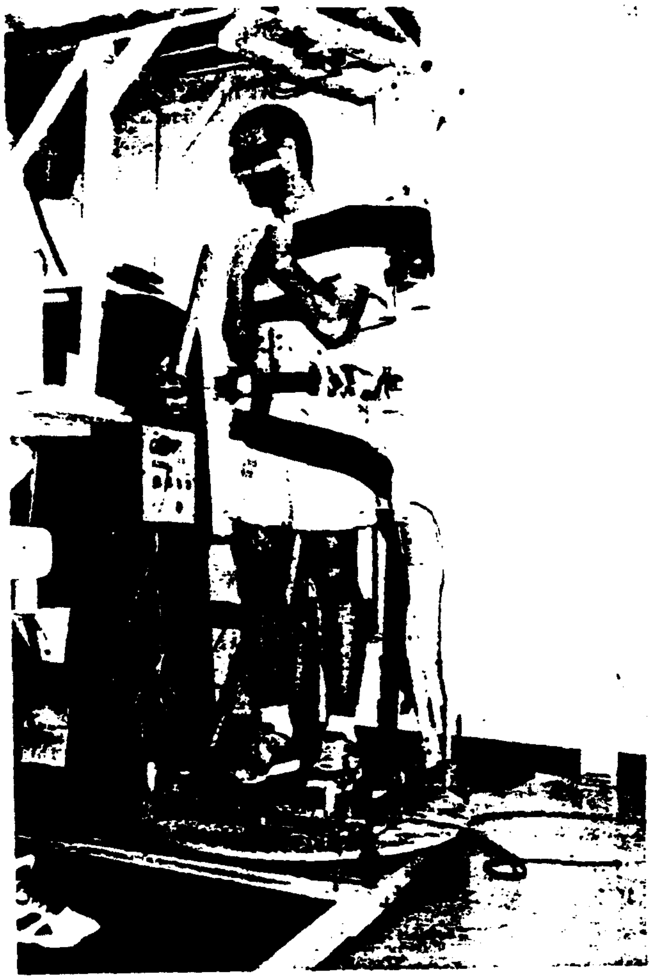

Figure 1. A twisting reference frame was employed to control and monitor static posture and dynamic motions of the subjects during torsional exertions.

Figure 1. Subjects were placed within a twisting reference frame and were asked to apply axial torque to a yoke that was placed around their back, shoulders, and chest. The yoke was connected to a Kin/Com isokinetic dynamometer (Chattanooga Group, Inc., Hixson, TX) whose motion axis was aligned vertically with the spine. The dynamometer provided a crude estimate of axial torque and also controlled the position and velocity of the twisting motion while restricting trunk motion to the transverse plane (i.e., twisting motion). A precise measure of three-dimensional trunk, reaction forces and moments was recorded from a force plate (Bertec 4060A; Bertec Corp., Worthington, OH) upon which the subject stood. A pelvic restraint limited twisting motion to the trunk while transferring three-dimensional trunk kinetics directly to the force plate. Position of the trunk was accurately measured during the experiment using a Lumbar Motion Monitor[22] that was attached at the subject's shoulder girdle and pelvis. A computer was used to graphically display the measured force plate torque in real-time so the subject could monitor the amount of torsional moment he was exerting. In addition to the subject's twisting torque, the monitor also displayed a target level and two tolerance lines that indicated the envelope of acceptable variation (±5%) from the designed level.

Electrical activity of the trunk muscles was collected using surface electromyography during each exertion. Bipolar surface electrodes were affixed to the skin over the muscles and connected to lightweight preamplifiers located near the electrodes. Signal passed through shielded cables to a hardware rack where they were further amplified and processed.[24]

All goniometer, force plate, and electromyograph signals were digitized at 100 Hz using an analog-to-digital converter and were recorded on a microcomputer. A separate microcomputer was used to control the dynamometer in the twisting reference frame.

Experimental Design. The experimental task consisted of four independent variables, including torsional moment, torsional direction, twisting position, and twisting velocity. Torsional exertions were performed at 100% and 50% of each subject's maximum voluntary contraction (MVC) effort. Exertions were performed in clockwise (CW) and counterclockwise (CCW) directions. Three twisting positions were observed. These were: 1) axial twisting while in the sagittally symmetric position, 2) twisting while rotated axially 20° to the right (20R), and 3) twisting while rotated axially 20° to the left (20L). Isokinetic exertions were performed at twisting velocities of 0 deg/sec (isometric) performed at the three twisting angles, and at 10 and 20 deg/sec over the range of 20R to 20L.

Normalized electromyograms of 10 trunk muscles served as dependent measures. The muscles sampled were the right and left pairs of the latissimus dorsi (LATR, LATL), erector spinae (ERSR, ERSL), rectus abdominis (RCAR, RCAL), external oblique (EXOR, EXOL), and the internal oblique (INOR, INOL). The sampling location for these muscles has been described previously.[34] The EMG signals were processed and normalized using activity levels collected during MVC exertions.

Procedure. Surface electrodes were placed over the muscles of interest using standard application procedures (National Institute for Occupational Safety and Health, 1991), and the quality of the signals were verified. Maximum integrated EMG values for the muscles of the trunk were established via MVC exertions. Because the isometric and isokinetic test exertions were performed in an upright standing posture, maximum EMG activity was collected during MVC exertions using the same posture.[33] To obtain the maximum EMG levels for those muscles that run primarily in the vertical direction (erector spinae and rectus abdomini), the maximal flexion and extension exertions were performed by the subject while in the twisting reference frame. Similar exertions were performed in the clockwise and counterclockwise twisting directions and in the right and left lateral directions to achieve maximum EMG levels from the oblique musculature (latissimus dorsi and internal and external obliques).

There was a rest period of 2 minutes between each trial to minimize the effects of fatigue. Maximum and submaximum isometric, torsional exertions were performed at each twisting angle. Isokinetic exertions were performed from a pre-rotated position of 24° through a symmetric posture to a final position of 24° on the opposite side. Clockwise and counterclockwise isokinetic exertions were collected at maximum and submaximum torsional levels. During submaximal exertions, if the subject failed to maintain the applied torque between the tolerance limits, the trial was repeated.

Analyses. The normalized EMG activities from the 10 trunk muscles were statistically analyzed to determine the level of coactivity between muscles and which muscles were responsible for changes in the experimental variables (i.e., force, velocity, direction, angle). Maximal trunk torque also was statistically described and analyzed to develop an appreciation for the magnitude of torques that could be exerted by the trunk. Formal statistical analyses consisted of multivariate analysis of variance applied to the collective muscle activities. This analysis was followed by univariate analysis of variance that evaluated the statistical significance associated with the individual activities of each muscle. The specific differences associated with the statistically significant results were determined via Tukey post-hoc analyses.

Twisting Model. The EMG-assisted model was exercised and results were generated from 320 individual trials to test the model validity and examine trunk loading under various static and dynamic twisting tasks just described for the experiment. Ten channels of integrated myoelectric data are used by the model to calculate relative muscle activities and account for the biomechanical influences of coactivity. Muscle tensile forces are determined by appropriately scaling the muscle forces and solving equations of dynamic equilibrium describing measured trunk moments. Three-dimensional spinal loads are determined from the vector summation of muscle force vectors.

Model Assumptions. The goal of the model was to biomechanically describe muscle activities and forces imposed upon the spine during torso twisting and torsional loading. Our objective was to develop a model that was realistic, validated, and accurate, while avoiding as much unnecessary complexity as reasonably possible. In this model, 10 muscle equivalent vectors approximate trunk anatomy and mechanics.[10,42] Trunk kinetics are described by active muscle forces without consideration of passive muscle, ligament, and disc moments.

The model assumes a linear relation between integrated EMG activity and muscle force. Whether EMG activity is linearly or nonlinearly related to joint torque has been debated in the literature,[16,17,35,47,49,51] but Hof and Van Den Berg[14] demonstrated EMG activity is linearly related to muscle force, whereas coactivation generates a nonlinear relation between EMG activity and joint torque. Because the EMG-assisted model accounts for muscle coactivity, a linear relationship was assumed.

Trunk Mechanics. Compared to former EMG-assisted models, this model is unique because multi-dimensional trunk moments and spinal loads are determined from dynamic muscle force vectors and moment arms. The mechanical orientation of the model is shown graphically in Figure 2. Muscle forces are represented as vector quantities between their two end points. Muscle origins are assigned a three-dimensional location relative to the spinal axis, representing the head of the force vector coplanar with the iliac crest (Table 1). Muscle insertions represent the tail of a force vector coplanar with the 12th rib. Using this approach, muscle orientations and lengths change throughout a movement, accounting for a muscle's changing mechanical advantage throughout the task. Dynamic computation of muscle lengths and vector directions permits muscle geometries to change as subjects rotate their trunks. In

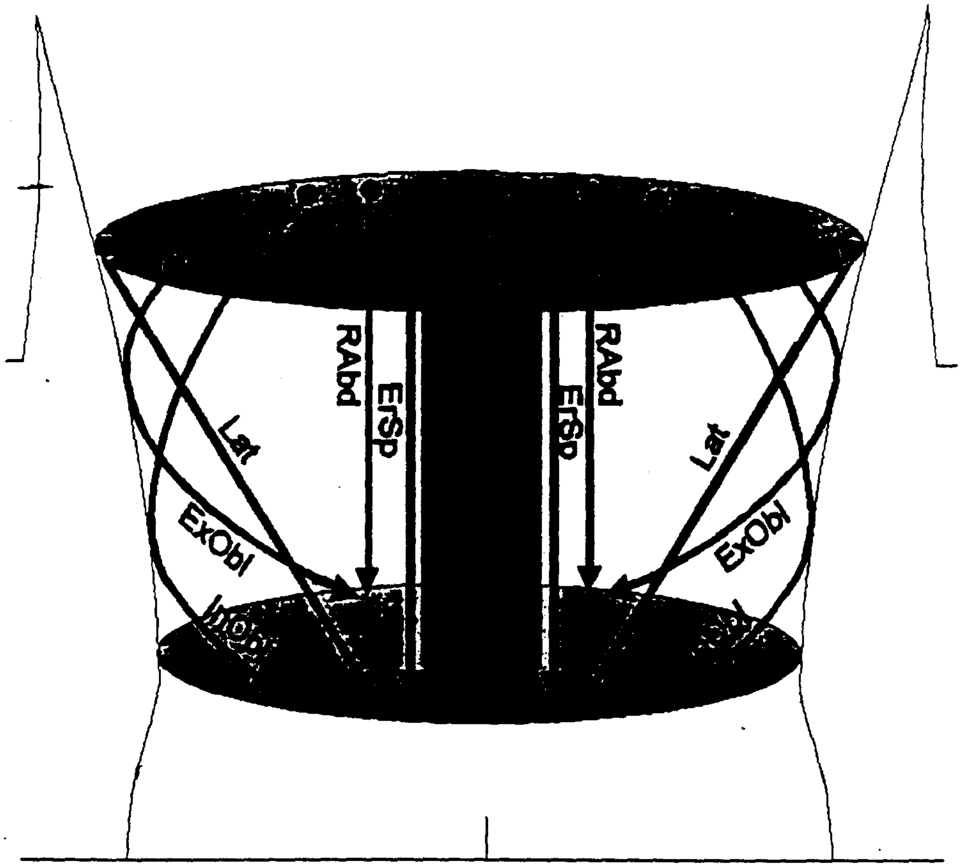

Figure 2. Trunk muscles are modeled as vectors determined from three-dimensional locations of muscle origins and insertions. Vector directions, lengths, and velocities are computed as a function of the instantaneous positions of the end points that move with the trunk posture.

addition, such a representation affords the opportunity to change the relative angles between the iliac and thoracic planes. Thus, pelvic tilt or asymmetric bends of the trunk could be simulated.

Spinal loading (compression, right-lateral shear, and anterior shear forces) is calculated from the vector sum of muscle forces. Muscle-generated moments about the spinal axis are predicted from the sum vector products combining dynamic tensile forces of each muscle, j, and respective moment arms.

$$\vec{M} = \sum_{i} \vec{r_i} \times \vec{F_i} \qquad (1)$$

Measured and predicted trunk moments are compared and must agree if the model is correctly simulating trunk mechanics. Statistical correlations between predicted and measured moment profiles serve as the measure of model performance and indicate how well the model accounts for the variability in the torsional moment. A high correlation implies the model generates an accurate simulation of dynamic spinal loading.

Muscle Force. Tensile forces generated by muscle equivalents are modeled as the product of normalized EMG activity, muscle cross-sectional area, muscle force per unit area (i.e., gain), and modulation factors that describe muscle behavior as

Table 1. Coefficients of Muscle Coordinates and Cross-sectional Areas

		Origins			Insertions		
Muscle Equivalent	Area	X	Y	Z	X	Y	Z
Right latissimus dorsi	0.0351	.25	−.30	0	.60	.10	0.0275 × Ht − 0.30 × Y
Left latissimus dorsi	0.0351	−.25	−.30	0	−.60	.10	0.0275 × Ht − 0.30 × Y
Right erector spinae	0.0389	.20	−.30	0	.30	−.30	0.0275 × Ht − 0.30 × Y
Left erector spinae	0.0389	−.20	−.30	0	−.30	−.30	0.0275 × Ht − 0.30 × Y
Right rectus abdominis	0.006	.10	.55	0	.10	.55	0.0275 × Ht − 0.30 × Y
Left rectus abdominis	0.006	−.10	.55	0	−.10	.55	0.0275 × Ht − 0.30 × Y
Right external abdominis oblique	0.0207	.10	.55	0	.45	−.19	0.0275 × Ht − 0.30 × Y
Left external abdominis oblique	0.0207	−.10	.55	0	−.45	−.19	0.0275 × Ht − 0.30 × Y
Right internal abdominis oblique	0.0215	.45	−.30	0	.45	.20	0.0275 × Ht − 0.30 × Y
Left internal abdominis oblique	0.0215	−.45	−.30	0	−.45	.20	0.0275 × Ht − 0.30 × Y

Area = coefficient × depth × width.
Anterior moment arm: X = coefficient × depth.
Right lateral moment arm: Y = [illegible]icient × width.
Elevation moment arm: Z.

a function of length (f[length]) and velocity (f[vel]).[10] This general relationship is governed by Equation 2.

$$\text{Force} = \text{Gain} \cdot \frac{\text{EMG(t)}}{\text{EMG}_{\text{Max}}} \cdot \text{Area} \cdot \text{f(Vel)} \cdot \text{f(Length)} \quad (2)$$

Time-dependent myoelectric data (EMG[t]) are normalized relative to maximum levels (EMG_{Max}) collected during maximum isometric voluntary exertions performed in three dimensions. Normalized EMG activity represents the fraction of maximum muscle activity that is applied at any point in time. The data allow relative comparison of activity from different muscles despite possible inter-electrode variability. Processed EMG data were smoothed within the model via a Hanning-weighted, time-domain 10-Hz equivalent-noise-bandwidth filter.[18]

The product of gain and area represents the force generating potential of a muscle equivalent. Another unique feature of this model is that muscle area must represent the maximum physiologic area of a muscle (Table 1) so that the force relation represents the fraction of maximum, possible tensile force. Maximum physiologic cross-sectional areas are employed in the torsional model as determined from subject anthropometry based on Schultz et al,[42] Dumas et al,[5] and McGill et al.[32]

Tensile force in skeletal muscle is influenced by relative muscle length and contraction velocity. Dynamic muscle lengths and contraction velocities of each muscle are determined from the instantaneous, three-dimensional locations of muscle origins and insertions based upon trunk position and subject anthropometry.[11,42] Unitless empirical relations that allow modulation of the equivalent muscle forces have been developed based on model output[9] and are shown in Equations 3 and 4.

$$\begin{aligned} f(\text{Length}_j) = {} & -3.2 + 10.2 \cdot \text{Length}_j - \\ & 10.4 \cdot \text{Length}_j^2 + 4.6 \cdot \text{Length}_j^3 \end{aligned} \quad (3)$$

$$f(\text{Vel}_j) = 1.2 - 0.99 \cdot \text{Vel}_j + 0.72 \cdot \text{Vel}_j^2 \quad (4)$$

Coefficients were determined by requiring muscle force per unit area to remain constant independent of muscle length and velocity. The empirically determined relations agree with the physiologic measurements and modulation factors proposed in the literature.[13,27,29] Vector directions of the muscle force equivalents are computed as straight line directions between each muscle's insertion and origin in three-dimensional space.

Muscle gain—i.e., force per unit area—is determined by solving the equation of dynamic equilibrium for torsional moment, and includes the influence of muscle coactivity as measured from EMG data. The physiologic literature has suggested that the gain must fall between 30 and 100 N/cm² to represent valid results.[30,41,48]

■ Results

Experimental Results

As expected, although subjects were asked to produce only torsional forces, significant forces about the other body axes were developed (Figure 3). Coupled moments generated in the sagittal plane were equivalent to 20% of the extension maximum and those generated in the coronal plane were equivalent to 79% of the lateral maximum during twisting exertions. Maximum torsional moment, and subsequent levels of 50% MVC, varied significantly ($P < .01$) as a function of twisting direction, rotation angle, and twisting velocity. Significantly more force was exerted in the clockwise direction (96 N·m MVC) compared with the counterclockwise direction (83 N·m MVC). In addition, significantly greater average torque was generated when the trunk was pre-rotated 20° away from the direction of applied torsion (107 N·m MVC) compared with the sagittally symmetric condition (95 N·m MVC). Likewise, greater torsional moments were generated during isometric exertions (52 N·m MVC) than during dynamic exertions (29 N·m MVC). Hence, these results show torsional moment is a function of twisting direction, angle, and velocity.

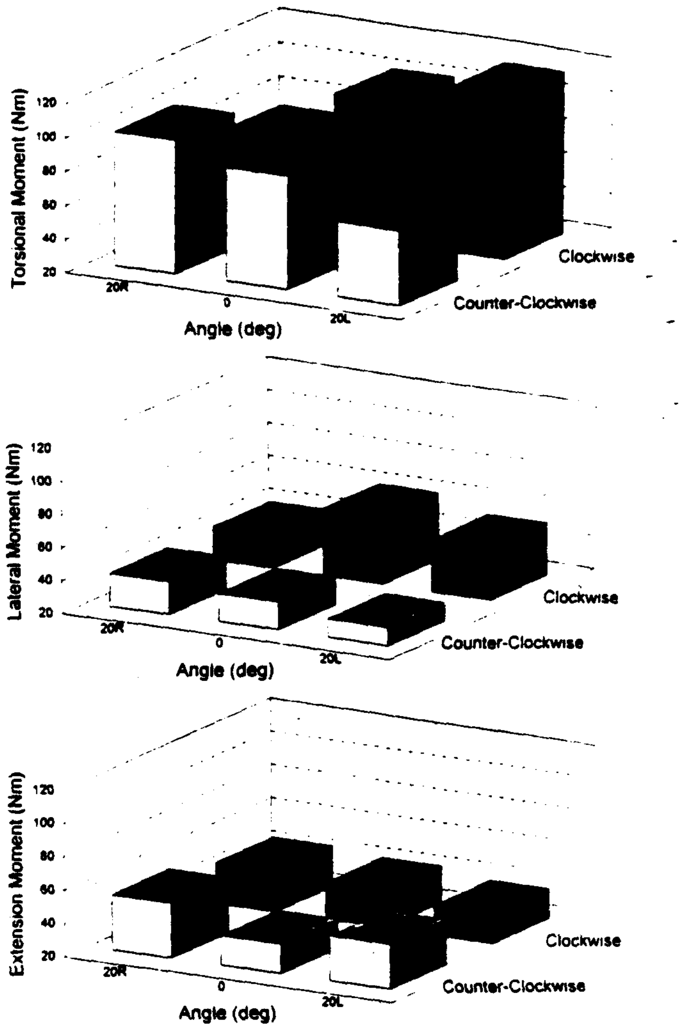

Figure 3. Maximum applied torsional moment is influenced by trunk pre-rotation and direction of twisting effort. Significant coupling moments—i.e., 20% extension MVC and 79% lateral MVC—were generated during torsional exertions.

The multivariate analysis of variance and analysis of variance results (Table 2) of relative EMG activities per unit axial torque (EMG-cost) indicated that force level, direction of applied moment, and twisting velocity in-

Table 2. Statistical Analyses Indicating Muscle Activities per Unit Torsional Moment

	Muscle	Force	Director	Angle	Velocity	Force × Direction	Force × Velocity	Dir × Velocity	Dir × Angle
MANOVA	—	$P < .0001$*	$P < .0001$*	$P < .7952$	$P < .0001$*	—	—	—	—
ANOVA	LATR	$P < .0026$*	$P < .0001$*	—	$P < .0022$*	—	—	—	—
	LATL	$P < .0081$*	$P < .0001$*	—	$P < .0001$*	$P < .0280$	—	$P < .0318$	—
	ERSR	—	$P < .0001$*	—	$P < .0001$*	—	—	$P < .0538$	—
	ERSL	—	$P < .0001$*	—	$P < .0001$*	—	—	$P < .0001$*	—
	RCAR	$P < .0006$*	$P < .0631$	—	$P < .0091$*	—	$P < .0693$	—	—
	RCAL	$P < .0009$*	$P < .0024$*	—	$P < .0031$*	—	—	—	—
	EXOR	—	$P < .0001$*	—	$P < .0001$*	$P < .0822$	—	$P < .0621$	—
	EXOL	—	$P < .0058$*	—	$P < .0001$*	—	—	$P < .0056$*	$P < .0785$
	INOR	$P < .0192$	$P < .0001$*	—	$P < .0002$*	—	—	—	—
	INOL	$P < .0510$	$P < .0001$*	—	$P < .0001$*	$P < .0795$	—	$P < .0011$*	—

Muscle activities per unit torsional moment were significantly influenced by exertion level, torsional direction, and twisting velocity. Although the electromyograms were scaled by applied moment, the latissimus dorsi, rectus abdomini, and internal obliques varied with exertion level.
* Significant at $P < .01$.
Only significant interactions ($P < .1$) are noted.
MANOVA = multivariate analysis of variance. LATR, LATL = latissimus dorsi right, left. ERSR, ERSL = erector spinae right, left. RCAR, RCAL = rectus abdominis right, left. EXOR, EXOL = external oblique right, left. INOR, INOL = internal oblique right, left.

fluence the relative activity of many of the trunk muscles. Further evaluations (Table 3) presented several interesting findings. First, although the results have been normalized to account for the relative changes in the torque production, EMG-costs of the latissimus dorsi, rectus abdominis ($P < .01$), and internal oblique ($P < .05$) muscles increased with exertion level. This trend is shown in Figure 4, where MVC exertions are shown to elicit at least 25% greater relative EMG than the 50% MVC exertions. The increased activity may indicate increased coactivity as the twisting torque level increases.

In addition, of particular interest is that many of the vertically oriented muscles change their activities as a function of the applied torsional direction. Such changes would be expected in the obliquely oriented muscles but not in the vertically oriented muscles. Table 3 indicates that the agonist muscles are generally at a higher level of activity in the CCW direction of force application. This effect is likely the result of the behavior of the left external oblique muscle, which is significantly more active when performing as an antagonist (i.e., counterclockwise) than when acting as an agonist (i.e., clockwise; Figure 5). Finally, the analyses indicated that exertions performed at increased velocities required greater EMG-cost. This effect is particularly prominent for the oblique and latissimus dorsi muscle groups (Figure 6). These findings are useful in determining which effects must be considered in the biomechanical model and which can be ignored.

To assess coactivity of the muscles, EMG-cost data were normalized relative to the most active muscle. Thus, muscle coactivity was described as the average activity level of the cocontracting muscles relative to the prime mover—i.e., most active muscle. Figure 7 indicates that the average coactivity during twisting exertions (42%) was significantly ($P < .01$) greater than during lifting (extension) exertions (26%).[9] However,

Table 3. Average Electromyographic Activity per Unit Torsional Moment Shown As a Function of the Main Experimental Conditions

	Force (MVC)		Direction		Angle			Velocity		
Muscle	50%	100%	CCW	CW	−20	0	20	0	10	20
LATR	0.68	0.98*	0.48	1.11*	0.63	0.85	0.73	0.66	1.03	1.16*
LATL	0.75	1.18*	1.51*	0.35	0.69	0.99	0.85	0.69	1.28	1.69*
ERSR	0.45	0.54	0.23	0.73*	0.49	0.52	0.34	0.39	0.64	0.82*
ERSL	0.51	0.65	0.92*	0.21	0.35	0.62	0.51	0.38*	0.68	1.34
RCAR	0.35	0.65*	0.39	0.55	0.31	0.51	0.45	0.37	0.7	0.73*
RCAL	0.37	0.68*	0.63*	0.36	0.35	0.54	0.46	0.38*	0.77*	0.78*
EXOR	1.01	1.29	1.39*	0.85	0.78	1.24	0.98	0.83*	1.59*	2.11*
EXOL	0.97	1.21	1.17*	0.95	0.74	1.18	0.89	0.78	1.53	1.99*
INOR	0.61	0.84*	0.46	0.93*	0.61	0.75	0.6	0.59	0.93	1.04*
INOL	0.71	1.02*	1.36*	0.31	0.57	0.92	0.73	0.61	1.16	1.57*

* Significant differences at $P < .05$.
The right latissimus dorsi, erector spinae, and internal oblique muscles increased with and were the primary generators of clockwise, torsional moment. Electromyography-cost (see text) increased significantly with velocity in all of the measured muscles. Relative activity was greater during sagittally symmetric exertions than during pre-rotated exertions, but was not statistically signficant.
MVC = maximum voluntary contraction. CCW = counterclockwise. CW = clockwise. LATR, LATL = latissimus dorsi right, left. ERSR, ERSL = erector spinae right, left. RCAR, RCAL = rectus abdominis right, left. EXOR, EXOL = external oblique right, left. INOR, INOL = internal oblique right, left.

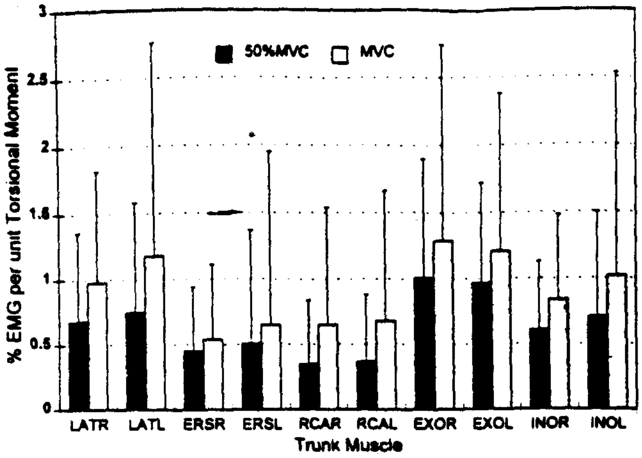

Figure 4. Despite the fact that the EMG activity was normalized per unit torsional moment, exertion level significantly influenced relative muscle activity. The influence is attributed to increased coactivity at higher levels of torsion.

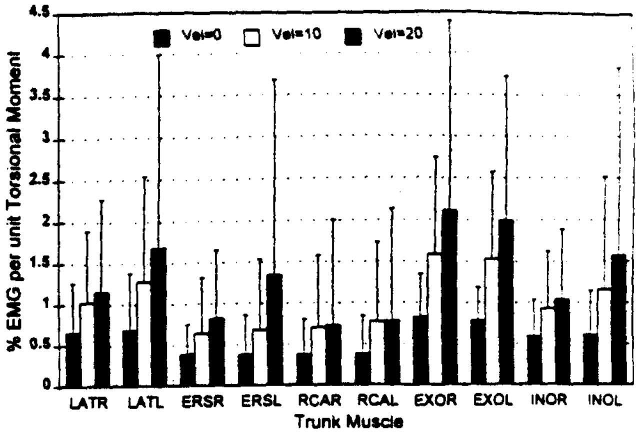

Figure 6. Electromyograph activity per unit torsional moment increased significantly with twisting velocity. The rise in coactivity was believed to occur in response to a greater required stability during dynamic exertions.

there were no significant differences in average coactivity between clockwise and counterclockwise exertions or twisting angle.

Model Results

Validation. Although there is no practical method for directly measuring multi-dimensional spinal loads *in vivo*, several model parameters may be employed to demonstrate the validity of the results. Predicted values of the calibrated subject muscle gains provide insight to the physiologic validity of the model performance. Subject gains averaged over all torsional exertions were 35.4 ± 23.4 N/cm^2 and represented a normal distribution. The average gain and distribution of values were well within the physiologically accepted range of 30 to 100 N/cm^2.

Examination of the modeled moment dynamics and variation permit the predicted results to be evaluated. The squared correlation coefficients (R^2) between measured and predicted torsional moments represent the accuracy of the modeled exertion. The model performed well during static and dynamic exertions. Distributions achieved from the 320 trials are shown in Figure 8 and illustrate the average R^2 was 0.80 and that more than 65% of the trials produced an R^2 greater than 0.8, and 36% greater than $R^2 = 0.9$.

The EMG-assisted model developed in this study was a good predictor of torsional trunk moments, as illustrated by the subject gain levels and the R^2 distributions. The relative tensile force in each trunk muscle was determined from measured myoelectric activity and was scaled via measured trunk moments to determine the magnitude of the force in each muscle. Because the equivalent muscle forces in the model are capable of

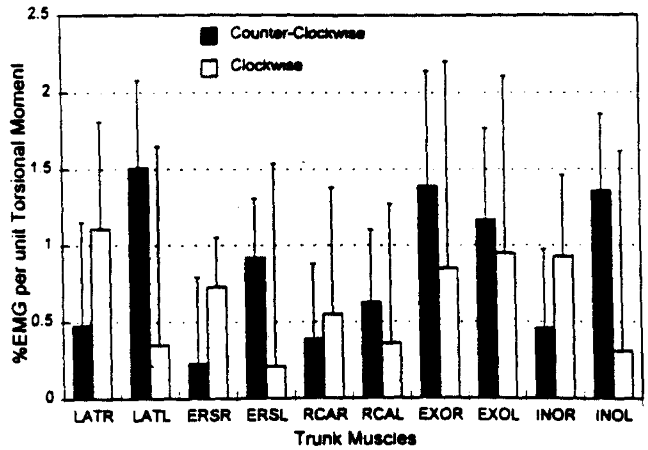

Figure 5. Electromyograph activity per unit torsional moment changed significantly with the direction of applied torque. Counterclockwise exertions generated greater overall activity levels, possibly due to coactivity. Note the left external oblique was more active when functioning as an antagonist than when it was functioning as an agonist.

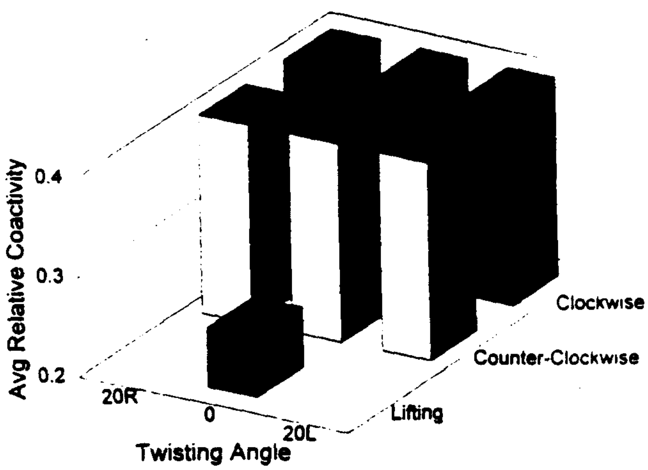

Figure 7. Average muscle coactivity during twisting exertions (42%) was significantly greater during lifting exertions (28%). There was no statistical difference between coactivity measured during torsional exertions as a function of twisting angle.

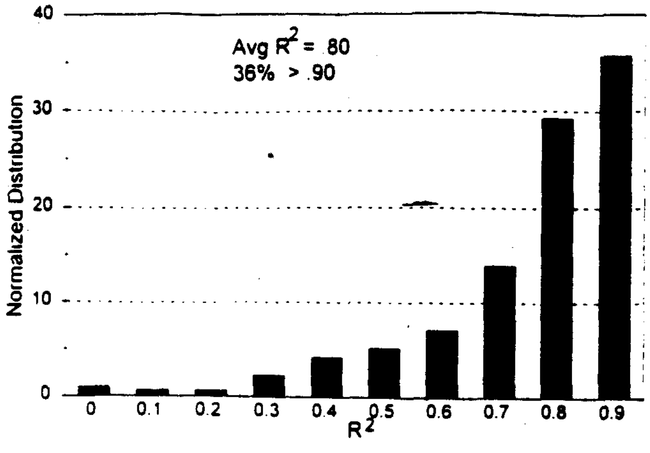

Figure 8. Squared correlations between measured and predicted torsional moments demonstrate the model accurately simulates dynamic as well as static twisting exertions.

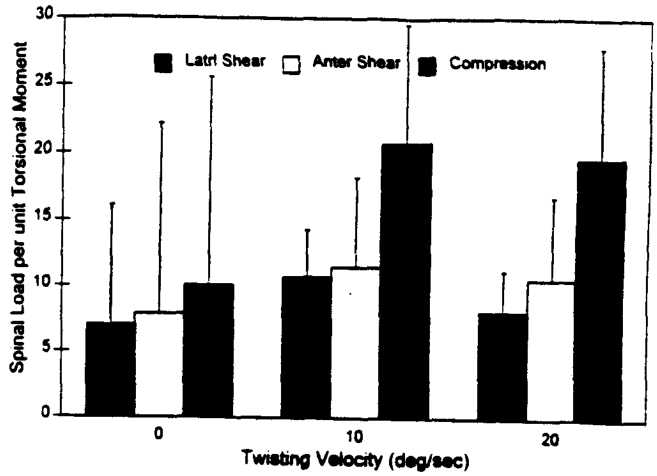

Figure 9. Spinal compression and shear per unit torsional moment are significantly influenced by twisting velocity.

accurately predicting time-dependent, dynamic and isometric twisting moments, the model was assumed capable of accurately simulating spinal loading.

Predicted Spine Loading. Spine loading, as predicted by the EMG-assisted model, was significantly affected by many of the variables manipulated in this experiment. Analysis of variance comparisons (Table 4) indicated that relative spinal loads (force per unit of torsional moment) changed ($P < .01$) as a function of exertion level, direction of the applied twisting torque, and twisting velocity. Figure 9 indicates increased twist velocity significantly increased spine forces in all three cardinal planes. However, it is particularly noteworthy that spine compression (per unit of twisting moment) magnitude doubled once twisting velocity occurred in the trunk. A pairwise comparison of relative spine load between the 20R and 20L conditions indicated that a significantly elevated compression load was present in the 20R condition (14.32 N/unit MHz) compared with the loading at the 20L condition (8.63 N/unit MHz). Thus, significantly different loadings were experienced in the spine depending upon whether the trunk was twisted to the right or the left. Finally, this analysis indicated that CCW motions resulted in about twice the anterior shear loading (10.18 N/unit MHz) compared with CW loading (5.37 N/unit MHz).

Table 4. Statistical Analyses of Spinal Load per Unit Torsional Moment

	Fx	Fy	Fz
Force	$P < .1559$	$P < .0071$*	$P < .0041$*
Direction	$P < .0001$*	$P < .1888$	$P < .6880$
Angle	$P < .0112$	$P < .2774$	$P < .1211$
Velocity	$P < .0008$*	$P < .1307$	$P < .0009$*

Results indicate that relative forces were influenced by applied moment, torsional direction, isometric twisting angle, and twisting velocity.
* Significant at $P < .01$.
No significant interactions at $P < .1$.

Discussion

Experimental Findings

This study has shown that the torso is limited in its ability to generate twisting torque, particularly under dynamic conditions. During MVC exertions, the 5-second average, torsional moments produced by our subjects were on the order of 52 N·m, and peak maximum moments were 90 N·m. Maximum isometric applied torsional moments measured during sagittally symmetric exertions (95 N·m) were similar to the moments reported by McGill[25,26] (91 N·m, 84 N·m clockwise, respectively) and Parnianpour et al[36] (96 N·m), although notably higher than reported by Parnianpour et al (66 N·m) in another study.[37] The twisting moments measured in the present study represent approximately 50% of the extension moment that subjects are able to produce, as reported by Parnianpour et al.[36] Hence, the body's ability to produce a twisting moment is far more limited than its ability to produce a lifting moment. We also have observed that the CCW moment production was less than 80% of the CW value. McGill[25] reported similar discrepancies between directions of applied moment. However, the trend in that study was exactly opposite ours. We believe this difference was simply the result of sign convention representing reactive versus applied moments in that study. Measured coupled moments indicated typical subjects are not capable of producing purely torsional moments without also generating extension (20% extension MVC) and lateral moments (79% lateral MVC).

Velocity also was critical in a subject's ability to apply a twisting moment. Once subjects were asked to move dynamically, even at low levels of velocity, average maximum torque production decreased by 43–50%.

When the true capacity to apply torsional moments, reduced torque capacity related to twisting velocity, and the effect of torque application direction are considered, it is not difficult to understand why twisting often is cited in the epidemiologic literature as a significant factor in occupationally related LBD. These analyses indicate that it is easy for task demands to exceed the capacity of the trunk in torsional exertions. This is particularly true when dynamic motion and CCW motion are involved.

These differences in moment generation capacity are an obvious reflection of the trunk muscle activities. Counterclockwise torque exertions may have been reduced because of increased coactive antagonism, most noteworthy in the left external abdominal oblique (Table 3), although the variation in average coactivity as a function of twisting direction was insignificant. Maximum torsion at pre-rotated angles was attributed to the relation between muscle length and contractile strength. Figure 3 demonstrates maximum torsion increases with length of the agonist muscles. McGill[27] agreed that muscle length may contribute to this effect, but concluded passive tissues may play a more significant role. Maximum torsional moments decrease with twisting velocity, as expected from the physiologic force velocity relation of skeletal muscle.[13]

We were able to address the issue of coactivation of the trunk muscles during twisting. In general, twisting results in about twice the amount of coactivation noted in trunk extension activities when normalized as a function of unit twisting moment.[9] Thus, more muscles are recruited at higher activation levels for twisting than for lifting exertions. The state-of-the-art has been unable to realistically predict muscle coactivity from biomechanical dynamics.[38] Electromyograph activity increased with torsional exertion in all of the measured trunk muscles—agonists and antagonists. Similar increases also were documented in EMG-cost as a function of exertion level (Table 3, Figure 4). Increased antagonist coactivity reduces the optimal nature of the applied force and cannot be predicated from biomechanical optimization functions that attempt to minimize spinal load, muscle force, or stress.[12,19]

Cursory examination of EMG data from the present study illustrated that clockwise torsional exertions were accompanied by increased activity in the right latissimus dorsi, right erector spinae, right internal oblique, and left external oblique muscles. The lines of action and points of origin and insertion of these muscles are situated so that each muscle generates a component of its moment vector in the clockwise direction about the spinal axis. Contralateral muscle activity generates counterclockwise axial moments. Evidence that the rectus abdomini are influenced significantly by force and not by torsional direction indicate these muscles are employed as trunk stabilizers, but are not recruited to contribute to the torsional moment. Pope et al[38,39] and McGill[25] also found large amounts of coactivity during axial twisting that they attributed to the need for trunk stability.

Because of the relationship between torsional moment and velocity, it could not be discerned whether variation in normalized EMG activity with respect to velocity was due to the change in velocity or the accompanying change in applied moment. Similar confounding can occur with applied moment and torsional direction as well as with moment and twisting angle. To remove this ambiguity, EMG activity per unit torsional moment (EMG-cost) was examined as a function of the dynamic, twisting parameters.

The EMG-cost increased with twisting velocity in all 10 of the measured muscles (Figure 6). This indicates the relative muscle activity of antagonist muscles increased as well as the prime movers. Increased coactivity has been described in the analyses of dynamic, sagittally symmetric, and asymmetric lifting exertions.[9,23,24] The phenomenon has been hypothesized as a method for increasing the system stability at higher velocities.[8,19,20,31] The results confirmed that static modeling cannot accurately represent dynamic lifting or twisting exertions.

The significant relationship between EMG-cost and exertion level (Table 2) is surprising in that EMG-cost has been used to intrinsically remove the influence of force from the analyses. Based on this, it must be concluded that the relationship between applied torsional moment and normalized EMG activity in the latissimus dorsi and rectus abdominis are nonlinear. This nonlinear behavior may be attributed to the coactivity and stabilizing behavior of the rectus abdominis. As already noted, the rectus abdomini do not appear to be recruited for their contribution to torsional moment, but are influenced by torsional force. Coactivity in that muscle may generate ancillary torsional moments, especially in pre-rotated postures. These coactive moments must be offset, possibly by latissimus dorsi activity, thereby creating a nonlinear behavior based upon the coactive stability forces in the rectus abdomini. The scenario permits a linear relationship between myoelectric activity and muscle force, while generating a nonlinear relationship between EMG activity and trunk moment.[14]

Model Predictions

A biomechanical model has been developed to determine trunk loading during torsional exertions. Optimization models have computed spinal loads from estimates of muscular activity during twisting exertions,[42] but have not yet successfully predicted muscular coactivity and antagonism. Analyses that incorrectly interpret muscle coactivity cannot accurately predict spinal loads. Electromyograph-assisted models incorporate the neuromuscular control system of the trunk musculature via measurement. Previous EMG-assisted models of torsional exertions[25,38] have predicted muscle forces that may exceed 250–700% of their capacity. The model developed in the present study predicts valid muscle forces employing measured coactivity, and accurately simulates torsional moments and spinal loads.

As indicated in Figure 8, this model accounts for an average of 80% of the variability in the measured torsional moments. Our previous models, which described lifting exertions, performed slightly better, with an average R^2 above .85. One of the possible reasons for this difference may involve some of the variance in body segment positions during some exertions. In particular, it was noted that some subjects flexed their head, which may have resulted in an unmeasurable change in trunk muscle lengths. However, we believe the magnitude of these differences were small, as evidenced by the model fidelity.

We believe the physiologically reasonable results generated by our model versus previous models can be attributed to three important differences in model design and assumption. First, our model included the activity and contribution of the right and left latissimus dorsi muscles, overlooked by some earlier studies.[38,42] Second, we used the maximum cross-sectional area of the muscle to predict the maximum force contribution of each trunk muscle.[25,32] Finally, our model treats the trunk musculature as a series of vectors capable of changing their orientation and mechanical advantage during a twisting motion.

These analyses have demonstrated that dynamic torsional trunk moments may be predicted from kinematic and EMG data within the constraints of physiologic validity. Compression and shear forces at the lumbosacral junction may be predicted from muscle equivalent forces scaled relative to the twisting moments. Therefore, qualitative, if not quantitative loading may be compared as a function of exertion load, trunk rotation, and motion characteristics.

The predicted spine loadings indicated that when any twisting velocity was present, significant increases in compression forces resulted (Figure 9). This velocity trend predicted by the model agrees with the dynamic relations proposed by Freivalds et al,[6] McGill and Norman,[28] Goel et al,[7] and Granata and Marras.[10] Those studies predicted increased compressive loading on the lumbar spine during dynamic exertions compared with isometric modeling.

In the present study, we also noted greater shear loading accompanying the higher compressive forces. Thus, twisting may create a situation where the combination of forces acting upon the spine place one at a higher risk of exceeding their tolerance to these forces.[43,44] These findings agree with the assessment of Marras et al,[21] who measured trunk motions associated with high- and low-risk lifting tasks. Statistically significantly greater velocities were observed in the high-risk jobs. The average velocity of these high-risk jobs was within 1.5 deg/sec of our 10 deg/sec velocity condition. Thus, the spine loading predicted by the model correlates well with epidemiologic findings.

Limitations

Several potential limitations must be considered when the results of the present study are examined. The dynamic geometry of the spinal curvature has been ignored in this study, and loads on individual spinal elements were not predicted. Further research could determine loads on individual elements of the lumbar spine by combining results generated by this model with a geometric model of the lumbar spine. Although the EMG length-modulation factor accounts for some of the passive muscle forces, passive spinal and ligamentous forces have been ignored in this analysis. The passive loads may play a role in spinal loading, especially during pre-rotated[38] and static[9] exertions.

■ Conclusions

We have gained a functional knowledge of how the musculoskeletal system behaves in twisting efforts. A model has been developed to accurately and validly predict dynamic torsional moments and simulate spinal loads from measured EMG data. Results from the model indicated that spinal load increased with exertion load, velocity, and twist angle. These results support epidemiologic findings that indicate risk of low back pain is related to exertion load, velocity, and twisting angle.[3,20,46] Future efforts must explore how the musculoskeletal system changes in more realistic, multidimensional, coupled postures and exertions.

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Address reprint requests to

W. S. Marras, PhD
Biodynamics Laboratory
The Ohio State University
1971 Neil Avenue
Columbus, OH 43210