

# Trunk Force Development during Static and Dynamic Lifts

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*Most lifting analyses have used static models to evaluate the loading of the trunk during the performance of work. Recent research has reported many differences in trunk muscle force capability when a wide range of trunk velocities are observed. This study focused upon these differences during slow trunk velocities, which would be expected during a lift. Forty-five subjects were tested for their ability to exert torque about their low back under static and dynamic sagittally symmetric lifting conditions. Trunk muscle electromyography was used as a measure of trunk loading. The results reveal that significantly greater loading occurs under slow dynamic conditions, as compared with static conditions. Hence, static models, which assume quasistatic motion, are not good indicators of dynamic trunk load. A regression model was developed that described trunk loading as a function of trunk angle and velocity. The correlation between torque and muscle electromyography in the erector spinae muscles was also found to increase significantly as trunk velocity increased. The implication of these findings upon lifting is discussed.*

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## INTRODUCTION

Ergonomists have long been concerned with the loading on the back that occurs during manual materials handling. Evidence in the literature indicates that low-back injury rates are correlated with the components of manual materials handling. Many researchers have found that, in industry, lifting is a task that increases the risk of low-back pain (Andersson, 1981; Davis, 1979; Svensson and Andersson, 1983). In response to these findings, ergonomists have tried to control the occurrence of low-back pain by designing workplaces and jobs in such a way as to eliminate this hazard from the task.

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Such a design process uses either biomechanical analysis techniques or physiological or psychophysical data to identify acceptable lifting conditions. The psychophysical and physiological evaluations do not provide specific quantitative information regarding the nature or magnitude of forces that are imposed upon the spine. Therefore, biomechanical analysis techniques based on strength evaluations are often used by those who design workplaces.

Biomechanical evaluations of lifting activities are limited to the analysis of the body in static positions. Several excellent models have been developed and validated that assess the loading within the trunk during such frozen lifting postures (Chaffin and Baker, 1970; National Institute for Occupational

Safety and Health, 1981; Schultz and Andersson, 1981). However, lifting and manual materials handling usually involve motion of the trunk. The proponents of many biomechanical techniques have tried to account for motion by making the assumption that a dynamic lift can be assessed piecewise by examining a series of static postures quasi-statically. However, the literature is devoid of information on how the motion component of a lift affects the load imposed on the trunk during that lift.

Marras, King, and Joynt (1984) compared the reactions of trunk load indicators during static trunk exertions and isokinetic trunk exertions. They found significant changes in the back-muscle forces occurring over a large range of back-motion velocities. Their study identified changes that occurred over a velocity range that was normalized according to a subject's maximum obtainable velocity. However, most of the changes that were observed fell outside of the velocity range that could be expected under normal lifting situations (Marras and Wongsam, 1986).

The objective of the present investigation was to study the static and slow dynamic trunk velocities that would occur during lifting. In order to achieve this objective, we defined the velocity levels in absolute terms (deg/s) over a small range of velocities. The loading of the trunk musculature was observed as a function of these velocities.

In many biomechanical analyses, the forces produced by the trunk muscles are considered as a measure of the loading of the spine. During a lifting activity, a moment is developed anterior to the spine. This moment must be counterbalanced by forces created by the trunk musculature. However, due to the proximity of the trunk muscles to the spine, this countermoment must be generated at a distance that is much less than that created by the lifting activity. Thus, the trunk muscles must create extremely large forces

(compared with the load). For example, if a 200 N (20.4 kg) box was held 61 cm from the spine, the back-muscle forces needed to simply hold the box (without moving) would exceed 2400 N. These trunk-muscle forces and the weight of the object lifted combine to create compressive, shear, and torsional forces on the spinal column. Hence, it is extremely important to accurately assess the activity of the trunk musculature when interpreting the loading on the spine. Additionally, the activity of the trunk muscles is directly proportional to the compressive loading that occurs at the spine (Schultz, Andersson, Ortengren, Nachemson, and Hader-speck, 1982a). Thus, the monitoring of trunk muscle activities provides a means by which to interpret the loading that occurs at the spinal column.

A measurement technique that defines the load experienced by a back muscle and which is thus closely related to the magnitude of the load experienced by the spine is *trunk muscle electromyography* (EMG). The source of the EMG signal is the electrical activity generated by the depolarization of the muscle-cell membrane during contraction. The integrated EMG has been recognized as a measure of force for the special cases of isometric exertions (Inman, Ralstan, Saunders, Feinstein, and Wright, 1952) or isotonic contractions at constant velocity (Bigland and Lippold, 1954). Specifically, the relationship between EMG activity, back muscle force, and spine load was found to be rather linear (Andersson, Herberts, and Ortengren, 1976; Andersson, Ortengren, and Herberts, 1977; Schultz et al., 1982a, 1982b).

In the present experiment, the point of rotation of the spine in the sagittal plane has been identified at the lumbro-sacral (L5/S1) junction (Cailliet, 1978). This evaluation assesses the activity of the back musculature, and thus the loading of the spine, while subjects exerted force about the spine similar to

that exerted during lifting. These exertions took place under both static and controlled dynamic conditions.

## METHOD

### Subjects

Subjects for this experiment were 45 healthy males. All subjects reported that they did not suffer from chronic low-back disorders. Subject age varied from 17 to 61 years, with a mean age of 31 years. A mix of occupations was represented in this population, including college students, faculty, and laborers.

All subjects were informed about the nature of the experiment prior to participation. They also had an opportunity to become familiar with the experimental apparatus before the experimental trials were conducted.

### Design

The independent variables in this study were trunk angle and trunk angular velocity. Trunk angle was defined at three levels (0 deg, 22.5 deg, and 45 deg), with the 0-deg angle corresponding to a normal standing posture. The 22.5- and 45-deg angles relate to forward flexion angles of the trunk relative to

the 0-deg upright posture. The velocity variable levels were defined in terms of trunk angular velocity. Four fixed velocity levels were defined (0 deg/s, 15 deg/s, 30 deg/s, and 90 deg/s). The 0-deg/s condition represents an isometric or static exertion. The other velocity conditions consisted of isokinetic angular velocities. Data were collected for each subject as a function of each velocity by angle condition. Thus, a repeated measures design was employed, in which the velocity and angle variables were completely crossed.

The following five dependent variables were defined in this experiment: (1) the maximum voluntary torque that a subject could exert about the L5/S1 junction during the return from a flexed trunk posture (this moment is similar to that experienced by the back link during the backlift), (2) the integrated EMG of the right latissimus dorsi muscle (LATR), (3) the integrated EMG of the left latissimus dorsi muscle (LATL), (4) the integrated EMG of the right erector spinae muscle (ERSR), and (5) the integrated EMG of the left erector spinae muscle (ERSL).

### Apparatus

The configuration of the experimental apparatus is shown in Figure 1. A frame of refer-

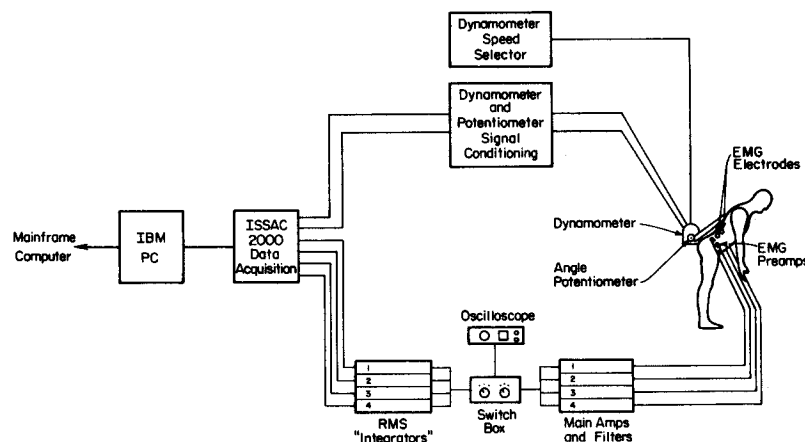


Figure 1. The experimental apparatus.

ence was used to orient the subject relative to the dynamometer. This reference frame was oriented in such a way that the torque about the L5/S1 junction could be measured. The reference frame also ensured that a constant-length lever arm was available against which the subject could apply force.

A Cybex II isokinetic dynamometer was used in this experiment. It permitted the preselection of the maximum allowable velocity (isometric or isokinetic) and trunk angle during the isometric exertion. An angular potentiometer attached to the dynamometer axis arm recorded angular position. Hip and leg straps were also attached to the reference frame, which limited trunk motion to that about the L5/S1 junction.

EMG signals were recorded with recessed surface electrodes placed over the muscle of interest. Skin impedance measurements and muscle site verification were performed prior to the experiment. The EMG signal was amplified at the muscle site using small, lightweight preamplifiers. This process reduced EMG noise dramatically. The signals were further amplified and filtered at a remote location (from the exertion area) according to standard procedures (Basmajian, 1978). At this point the signals were visually examined using a switchbox and oscilloscope. This check was performed prior to each experimental condition to ensure data quality. Each EMG signal was then passed through an "integrator," where the signal was rectified and averaged via a root mean square (RMS) procedure.

The EMG, dynamometer, and potentiometer signals were monitored on-line by an ISAAC 2000 data acquisition system. This system was in turn monitored by a microcomputer, where the data were stored and conditioned. Advanced statistical analyses were performed on the university's mainframe computer.

#### *Experimental Task*

Once the subject was strapped into the experimental reference frame, he was required to produce a maximal voluntary torque against the axis arm of the dynamometer. The exertion started with the trunk at a 60-deg forward angle and culminated past the 0-deg angle for isokinetic exertions. The first 15 deg of motion permitted the subject to attain an isokinetic velocity state.

Isometric exertions were performed in a similar manner, except that the trunk angle was preselected and no motion was allowed. Both isokinetic and isometric exertions were sagittally symmetric and required the subject to exert maximum voluntary torque against the dynamometer throughout the entire range of motion.

In this manner, the tasks were able to simulate the loadings experienced by the back with and without motion involvement.

#### *Data Treatment*

In order to assist in the data analyses, hard-copy plots of the pretest (rest calibration) and of each exertion were produced. These plots showed the activity of all dependent variables on a common time scale. A hard-copy plot was produced for each experimental condition and for each subject.

Statistical data were collected for each signal via a computer program run on the microcomputer. This program determined the starting point, end point, maximum, minimum, mean, and variance for each dependent variable under each condition. The hard-copy plots were used to narrow the search region for the start and end times, thus reducing processing time.

A program was also employed to normalize the data as a function of the experimental conditions for each subject and each muscle. This program first examined the intensity of

activity at the window (0.12 s) of interest (angle-velocity combination). Next, the rest level corresponding to the angle was subtracted from the window intensity. The maximum activity of each signal was then found, and the resting signal was subtracted from this maximum. Finally, the window (minus rest) was divided by the maximum signal (minus rest). This normalizing procedure, used at each velocity-angle combination (V, A), is shown in Equation 1.

Normalized Signal (V,A) =

$$\frac{\text{window value (V,A)} - \text{rest (V,A)}}{\text{maximum value} - \text{rest (V,A)}} \quad (1)$$

Once all signals had been normalized (torque signals were also analyzed in non-normalized form), they were compiled into a file and evaluated with advanced statistical analysis techniques.

## RESULTS

### Muscle Force Characteristics

Univariate analysis of variance was used to test the significance of each dependent variable. Table 1 summarizes the results of these tests. This table indicates that each variable responded significantly to changes in velocity, angle, and the Velocity  $\times$  Angle interaction.

TABLE 1

ANOVA Summary of Dependent Variables<sup>1</sup>

	<i>Latissimus Dorsi</i>		<i>Erector Spinae</i>		
	<i>Right</i>	<i>Left</i>	<i>Right</i>	<i>Left</i>	<i>Torque</i>
Velocity	16.91**	24.82**	46.74**	82.10**	143.87**
Angle	3.42*	3.86*	133.51**	136.81**	4.40**
V $\times$ A	2.96**	3.01**	5.63*	6.94**	3.19**

<sup>1</sup> Values indicate the univariate *F* value.

\* Significant at  $p < 0.05$

\*\* Significant at  $p < 0.01$

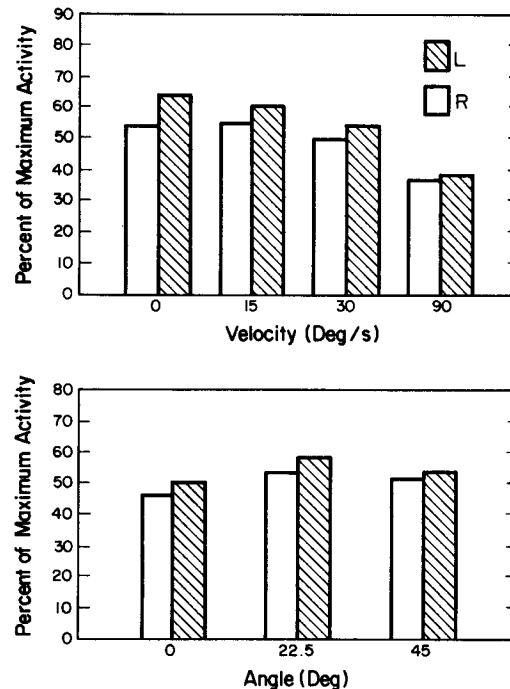


Figure 2. Marginal mean electromyographic activity of the latissimus dorsi right and left muscles as a function of velocity and angle.

The activity of the right and left latissimus dorsi muscles is shown in Figure 2. This figure indicates the mean relative force that is produced in the muscles as a function of the marginal angle and velocity conditions. It also indicates that the latissimus dorsi

muscles decrease their activity as the velocity condition increases.

Post hoc analyses were performed to determine which velocity and angle conditions contributed to the univariate significance of each dependent measure. The results of Duncan Range Tests are shown in Figure 3 for all dependent measures. Generally, these results indicate that the latissimus dorsi muscle forces were distinguishable between the 30 deg/s, 90 deg/s, and all other velocity conditions. Reactions of these muscles were

		Velocity (Deg/s)			
		0	15	30	90
Velocity (Deg/s)	0		ER EL TT	LR LL ER EL TT	LR LL ER EL TT
	15			LL ER EL TT	LR LL ER EL TT
	30				LR LL ER EL TT
	90				

		Angle (Deg)		
		0	22.5	45
Angle (Deg)	0		LR LL ER EL TT	ER EL
	22.5			ER EL TT
	45			

Figure 3. Post hoc (Duncan Range Test) analysis of experimental conditions for all dependent measures. A symbol in a cell indicates that there is a significant difference between conditions for the dependent measure. LL = latissimus dorsi left, LR = latissimus dorsi right, EL = erector spinae left, ER = erector spinae right, TT = trunk torque.

indistinguishable between the static and 15 deg/s conditions. This figure also indicates that within the latissimus dorsi muscles the only significant changes in muscle force as a function of angle occurred when the 0-deg angle was compared with the 22.5-deg angle.

A difference in response levels between the right and left latissimus dorsi muscles is also apparent. The left muscle generally must produce between 1 and 10% greater relative force in order to perform this sagittally symmetric task. However, the difference in activity depends on the velocity condition. Under static conditions the difference is the greatest (10%), but, as the velocity conditions increase, the difference between conditions decreases monotonically. Under the 15 and 30 deg/s conditions, the difference between the muscle force is 5% and 4%, respectively. Under the 90 deg/s condition, however, the difference is only 1%.

The response of the right and left erector spinae muscles to the marginal velocity and angle conditions is shown in Figure 4. This set of muscles responded very differently to the experimental conditions than did the latissimus dorsi muscles. This figure indicates that the 15 deg/s condition elicits the greatest amount of force from the erector spinae muscles, followed by the 30 deg/s condition. The static condition produces the third-greatest amount of muscle force and the 90 deg/s condition produces the least muscle force. Post hoc analyses of the erector spinae responses are also shown in Figure 3. This analysis indicated that, generally, every velocity and angle condition produced a significantly different response. Unlike the latissimus dorsi muscles, the right and left sides of this set of muscles produced similar responses.

#### Trunk Torque

As shown in Figure 5, the torque capacity of the back decreases as the velocity in-

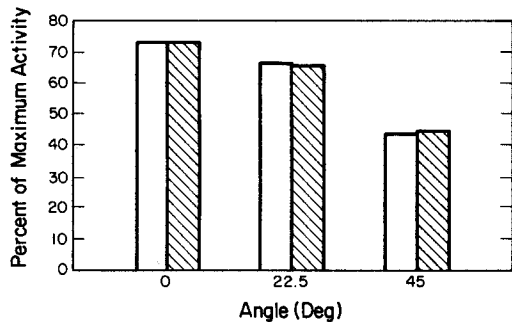
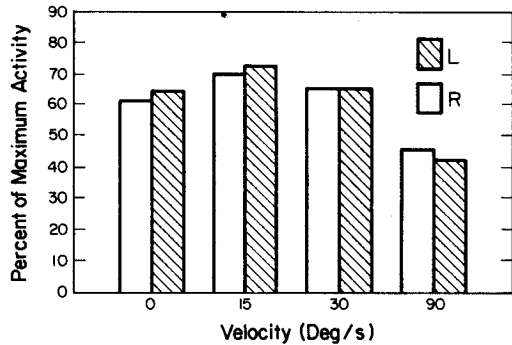


Figure 4. Marginal mean electromyographic activity of the erector spinae right and left muscles as a function of velocity and angle.

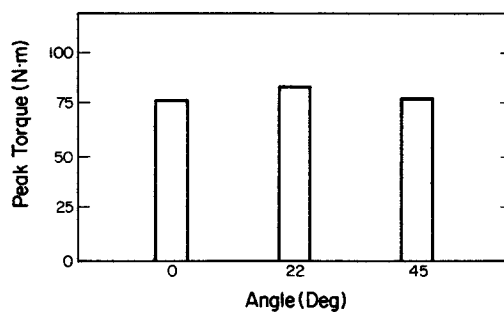
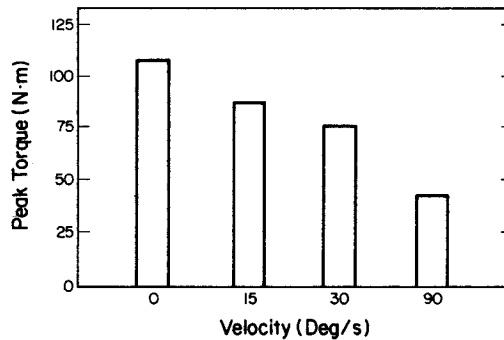


Figure 5. Absolute torque activity as a function of velocity and angle.

creases. Post hoc analyses (Figure 3) indicated that the torque variable responded differently to each velocity condition. Analyses of the angle conditions indicated that the 0-deg and 45-deg conditions yielded identical torque responses of the back, whereas the 22.5-deg condition yielded torque responses that were significantly greater (about 13.5 Nm).

This study also evaluated the ability to predict dynamic back strength from static trunk-strength measures. The correlation between torque capabilities among the various experimental conditions is shown in Table 2, which indicates that most conditions were positively correlated. However, the experimental conditions, particularly the velocity condition, affected the strength of the correlation. Strength was well-correlated with most velocity conditions. However, the

strength of the correlations decreased as the differences between velocities increased. As indicated in Table 2, static torque capability was not a good predictor of dynamic torque capability.

The correlations between the muscle activities and trunk torque were also investigated. Significant differences in these correlations were noted, as indicated in Figure 6. The correlation between the latissimus dorsi muscles remains rather stable as the velocity condition increases. However, the relationship between the erector spinae muscles and torque increases dramatically as the velocity condition increases.

*Predictions of Trunk Loadings*

As discussed earlier, the force production activity of the trunk muscles is often used as a measure of trunk loadings. In order to as-

TABLE 2

Back Strength (Torque) Correlation Matrix

		SO	22	45	D 15 0	D 15 22	D 15 45	D 30 0	D 30 22	D 30 45	D 90 0	D 90 22	D 90 45	
S	0	1												
S	22	0.76	1											
S	45	0.66	0.84	1										
D	15	0	0.65	0.65	0.65	1								
D	15	22	0.64	0.66	0.67	0.86	1							
D	15	45	0.45	0.60	0.61	0.45	0.66	1						
D	30	0	0.69	0.62	0.60	0.74	0.69	0.47	1					
D	30	22	0.66	0.66	0.59	0.72	0.82	0.64	0.86	1				
D	30	45	0.51	0.60	0.50	0.52	0.62	0.64	0.59	0.73	1			
D	90	0	0.38	0.45	0.43	0.36	0.39	0.44	0.58	0.52	0.51	1		
D	90	22	0.38	0.50	0.49	0.45	0.44	0.53	0.56	0.57	0.60	0.94	1	
D	90	45	0.31	0.46	0.47	—	—	0.42	0.34	0.36	0.56	0.71	0.82	1

Significance = 0.05

S = static angle (deg)

D = dynamic velocity (deg/s) at a given angle (deg)

sess the loading due to motion and angle, prediction equations were developed that describe the effects of the synergistic activity contributions of all four back muscles.

Models were tested that predicted the trunk loading (based on back muscle activi-

ties) as a function of the various velocity and angle characteristics of the trunk. The equations that represent the experimental data most accurately are shown in Table 3. These equations display the best normality characteristics. The regression equations indicate that between about 33% and 60% of the variance can be explained by considering only the relationships between velocity and angle conditions. The polynomial regression allows one to explain an additional 25% of the back loading variability over the linear prediction equations.

## DISCUSSION

The results of this study provide several types of evidence that static loading of the trunk varies markedly from loadings that are experienced during motion, as is the case during lifting. This research has quantitatively described these differences. The velocity and angle conditions yielded statistically significant reactions from all of the dependent variables. The effects of these reactions on the loading of the spine will be discussed here.

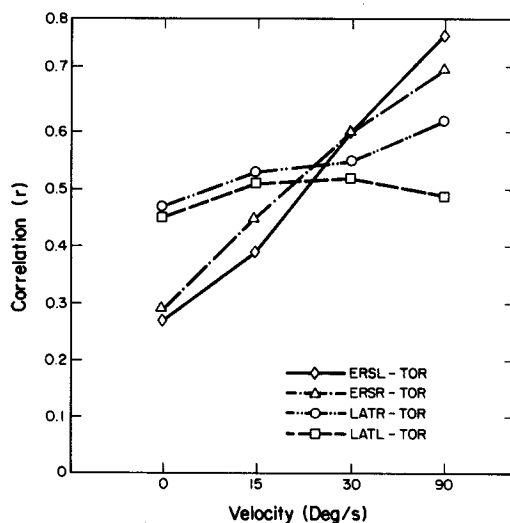


Figure 6. Correlation between muscle EMG and torque (TOR) as a function of velocity.



TABLE 3

Predictions of Load on the Spine (Measured by Back Muscle Activity) due to Changes in Trunk Angle and Velocity

Highest Equation Component	Regression Equation	R <sup>2</sup>	Normality
Linear	Load = 3.122985 - 0.010536V - 0.018797 A	0.3311	0.99176
Polynomial	Load = 0.296391 + 0.000731251V - 0.000610522 A - 0.000238514V <sup>2</sup> - 0.0000529227A <sup>2</sup>	0.5864	0.99503

V = trunk velocity  
A = trunk angle

The latissimus dorsi muscles generally decrease their activity as velocity increases and do so in a linear fashion. Thus, less spine loading would occur due to the activity of these muscles when motion is introduced into a task. However, it must be kept in mind that the latissimus dorsi muscle has a fairly small cross-sectional area and has only a small mechanical advantage over the erector spinae muscles. Hence, the contribution of this muscle during maximum lifting is rather predictable as a function of velocity. A significant difference was noted in the activity between the left and right latissimus dorsi muscles as a function of velocity. The differences between the right and left sides of the body decrease as the trunk velocity increases. This may represent the fine-tuning control of which these muscles are capable under slow exertion conditions. The difference may also represent the greater strength that most people have on the dominant side of the body. Hence, since the right arm (the latissimus dorsi muscle innervates the right shoulder complex) is stronger, a smaller percentage of the maximum force is required to produce a sagittally symmetric exertion on the right side. As velocity increases, less precise balancing would be required since the subject simply exerts force as quickly as possible.

This muscle also responds uniquely to angle. The greatest activity occurs at the 22.5-deg trunk angle, indicating that, at this angle, the muscle has the greatest mechanical advantage.

The erector spinae muscles respond quite differently to motion. Due to their size and location, these muscles account for a large portion of trunk loading. The activity of these muscles on either side of the body was rather similar. Such a finding was not unexpected, since the erector spinae muscles are medially located, as compared with the latissimus dorsi muscles, and would not be as involved in the dominant-side control of the body.

The erector spinae muscles increase their activity under slow dynamic conditions as compared with static conditions. This finding is particularly significant when one considers that the latissimus dorsi activity decreases only slightly under these slow conditions, and the torque-supporting capability of the back decreases substantially. Hence, under slow lifting conditions, the loading of the spine increases substantially yet there is less lifting capacity, especially under the 15-deg/s and 30-deg/s conditions. Thus, even slow lifting may load the spine more than has been predicted previously based on static exertions.

The significance of these findings can be

appreciated by comparing the muscle activities with torque production under the various conditions. For example, the erector spinae activity can be compared with the torque activities shown in Figures 4 and 5. A comparison of the muscle-to-torque ratios under the various conditions shows that, compared with static conditions, an increase in muscle activity of over 40% is required to produce a given amount of trunk torque when the trunk is moving at 15 deg/s. The ratios increase to over 80% when the static conditions are compared with the 90-deg/s condition. Thus, static models of lifting that assume that slow lifts are quasi-static may severely underestimate the spine loading that occurs. These results indicate that the loading cost to the trunk changes in an unexpected manner when velocity is introduced into a lift.

The regression equations that were developed based on this data represent an estimate of the true cost of spine loading during maximal trunk exertions. The lack of a greater  $R^2$  would also indicate that the sudden increase in spine load during slow dynamic lifts was underrepresented in these linear approximations. Future research efforts should also focus on the relative costs (per unit torque) of trunk motion. The regression equations developed in the present research were able to account for a substantial amount of variability in the data. They might be even more predictive if anthropometric parameters, age, sex, motivation, and erector spinae nonlinearities were accounted for. Future research efforts should take such factors into account in the development of regression equations that explain trunk loading.

This research has also shown that electromyographic recordings of isokinetic exertions are well correlated with the amount of torque being exerted. In fact, the correlation between torque and EMG increases substantially as the trunk velocity increases, thus in-

dicating that monitoring the erector spinae EMG under motion conditions is a fairly accurate indicator of the external moment supported by the trunk. This finding may be used as a quantitative tool for workplace design under motion conditions.

Finally, the evaluation of trunk torque generation has shown that measures of static exertions of the trunk are not a good indication of dynamic trunk strength. When lifting tasks involving motion are of interest, dynamic (isokinetic) strength testing should be performed to assess the subject's capability instead of extrapolating capability from static test data (as is currently the practice). Furthermore, the trunk velocity should be matched to that which would be expected in the workplace.

#### ACKNOWLEDGMENTS

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