TRUNK MOTION DURING LIFTING: THE RELATIVE COST

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ABSTRACT

The cost of lifting motion to back loading has been investigated traditionally by monitoring the electromyographic (EMG) recordings of trunk muscle during a controlled lift. When subjects are tested on an isokinetic dynamometer, the EMG activity decreases as the velocity increases and trunk torque production also decreases with added velocity. However, during an actual lift, the necessary torque needed to handle the load remains constant regardless of the lift speed. This research has investigated the muscle force per unit torque which is needed to support a load under various trunk velocities. Forty-five subjects were tested for maximum torque production under various velocity and angle conditions. The relative trunk loading cost of velocity was evaluated and described in an equation for slow velocity (0–30 deg / s) trunk exertions. These results were used to discuss how static lifting models might be adjusted to account for the added trunk load due to velocity.

INTRODUCTION

The etiology of back disorders as related to lifting activities is not well understood. However, evaluation of the mechanical and biomechanical factors which come into play during a lifting task has provided a promising means to understand the problem. Biomechanics assumes that lifting capacity is a result of forces acting upon the spinal column. These forces are due to both *external loading* factors generated by the object being lifted and *internal loading* factors generated by muscles and pressures within the body. The internal loadings due to the muscle activity usually far exceed the forces due to the external load.

This is due to the fact that internal forces within the muscles must be used to balance the external load. However, the internal force must act at a biomechanical disadvantage since the moment arm between the muscle and the spine is at a much shorter distance compared to the distance between the external load and the spine. Hence, monitoring of the internal forces is often a good indicator and is often proportional to the total load experienced by the trunk during a lift.

It is believed that when these trunk forces become excessive or awkward they stimulate nociceptors within the skin, joints, muscles and bones of the subject causing low back pain (LBP). Knowledge of the synergistic ef-

fects of these internal and external forces can be used to minimize the possibility of injury due to workplace design. The significance of biomechanics to low back research is evident from the common occurrence that once someone has experienced a low back problem his physical capabilities become limited. The person is not able to use his body and dynamically exert force about his environment as freely as he once did. Thus, the low back disorder is manifested by a change in the biomechanical capabilities.

Marras et al. (1984) found that these internal forces changed dramatically when static postures of the trunk were compared with dynamic (isokinetic) motion of the trunk. In particular they found that the latissimus dorsi and erector spinae muscles' activities changed dramatically when lifting positions were evaluated under static and motion conditions. These researchers evaluated four velocities in a subjects motion capability which ranged from static to their maximum possible velocity. All variables were studied relative to this maximum possible trunk velocity.

More recent studies by Marras and Wongsam (1986) have investigated the trunk velocity attained by normal subjects performing lifting tasks without any external loads. These studies found that most trunk velocities exhibited during lifting did not fall within the range investigated in the earlier study by Marras et al. (1984). Furthermore, it was found that most trunk velocities observed were fairly slow since lifters often accomplish much of the lifting motion through rotation of the pelvis about the femur rather than in the back. Therefore, an experiment was performed which investigated the action of the back muscle internal forces and external force exertion capability while subjects exerted force at slower velocities. These velocities were fixed and were not considered relative to individual velocity capabilities as was done in earlier studies. They were also compared with static trunk action. Hence, this experiment would

provide a data base of internal force action over the range of velocities employed during lifting.

This research effort investigated the action of internal and external trunk loading factors as a function of the trunk velocity. During the performance of a manual materials handling (MMH) task the external load remains relatively constant unless the lifter changes the distance of the load from the spine. Hoever, the trunk supporting characteristics (torque production) and the internal force activity are expected to change as a function of velocity. Therefore, in order to facilitate the usefulness of this research as applied to actual lifting situations, the internal trunk loading was investigated in relative terms. In other words, the amount of internal force within the muscles necessary to support a given amount of torque about the trunk was investigated. In this manner the effects of motion upon the spine could be isolated.

METHODS

Approach

The experimental focus in this research was limited to the motion about the lumbro-sacral junction (L5/S1). This junction is significant for several reasons. First, this junction is the site of a significant amount of motion in the spine. Caillet (1968) has reported that 80–90% of lumbar sagittal plane motion occurs at this point. Second, the posterior longitudinal ligament is narrowest at this point, thus, suggesting load bearing limitations. Finally, L5/S1 has been associated with LBP and has been identified as the weak link of the body during MMH.

The literature (Andersson, 1981; Davis, 1979; Strasser, 1980) has also suggested that MMH work involving maximum exertions in repetitive bending postures were hazardous. Therefore, this study has investigated trunk

posture and trunk motion effects while subjects are exerting maximum voluntary force.

In order to assess the effects of motion in this highly controlled experiment, several assumptions were made: First, the load experienced by the spine during a force exerting motion was assumed to be a function of the torque induced about the spine. In this study, the biomechanical action of the trunk was of interest. Therefore, we may assume that the torque about the spine link during a trunk exertion may be directly measured at the back and it is not necessary to use the arms (and require the subject to lift) to investigate the loading of the spine.

The next assumption is that we may determine the force exerted by a muscle, and thus internal structure trunk load, via integrated electromyography (EMG). Many researchers have documented the relationship between force and EMG under static conditions (Inman et al., 1954; Andersson et al., 1976; Andersson and Schultz, 1979). Other researchers have also shown relationships between EMG and muscle force under constant velocity (Bigland and Lippold, 1954). The present experiment has been designed so that the relative force produced by the muscle may be evaluated under motion conditions.

The final assumption provides a framework for the assessment of the "cost" of lumbar motion. It is assumed that the compressive forces of the trunk may be assessed instantaneously throughout a lift by using the equilibrium analysis techniques derived by Schultz and Andersson (1981). Here the forces acting along an imaginary transverse plane within the trunk are related by the six force and moment equations. We will presume that a state of equilibrium must also exist during isokinetic exertions of the trunk. Therefore, we may use trunk torque and muscle force inputs to analyze compression experienced by the spine.

Subjects

Subjects in this experiment consisted of 45 healthy males who had a negative history of significant low back pain. Subject ages varied from 17 to 61 years with an average age of 31 years and a standard deviation of 8.6 years. The mean height of subjects was 179.3 cm and the mean weight was 79.8 kg. A mix of occupations are represented including students, faculty and laborers.

Subjects were informed about the nature of the experiment prior to the test and participated in the experiment on a voluntary basis. Subjects were also given an opportunity to warm up and become familiar with the equipment.

Design

The independent variables in this experiment consisted of trunk angle and trunk angular velocity. Marras and Wongsam (1986) found that the mean maximum trunk angle during lifting was about 45 deg. Therefore, the angle variable was fixed at three equally spaced levels within this range. Trunk angles consisted of 0, 22.5 and 45 deg of forward bend from the upright (0 deg) position.

Marras and Wongsam also studied lifting motions of normal subjects and found that the mean maximum lifting velocity was about 80 deg/s with most normal lifting velocities occurring at much lower velocity levels. Therefore, the velocity range was defined between 90 deg/s and 0 deg/s (static) with a greater emphasis at the slower velocities. Hence, the velocity variable was fixed with four levels defined at 0, 15, 30 and 90 deg/s. The 0 deg/s velocity was a static (isometric) exertion of the trunk while all other velocity levels were isokinetic. .

The dependent variables consisted of the external and internal trunk loading factors. The external loading factor consists of the load supporting capability of the trunk during

a back lift task. This variable was measured at the back link and consisted of the maximum torque a subject can produce about L5/S1. The internal loading factors consisted of back muscle forces which have been observed to change with trunk velocity (Marras et al., 1984). The relative muscle force of the muscles was measured by monitoring the "integrated" (RMS) EMG signals of the muscle. Four muscle forces were monitored and consisted of: (1) the right latissimus dorsi muscle; (2) the left latissimus dorsi muscle; (3) the right erector spinae muscle; (4) the left erector spinae muscle.

The experimental design consisted of a repeated measure design where the velocity and angle factors were completely crossed. Each subject was tested once under each condition unless the subject indicated that he did not produce a maximum exertion. If a maximum exertion was not produced, the trial was repeated. Subjects were permitted at least a two minute rest period between trials.

Apparatus

The subject was placed in a reference frame during experimental exertions. This frame positioned the subject relative to the dynamometer so that a sagittally symmetric exertion could be performed. The reference frame and configuration of the dynamometer and EMG recording method was similar to that described by Marras et al. (1984) with several exceptions. First, the EMG signals were "integrated" with hardware techniques after they were passed from the switch box but before they were recorded. Next, all signals were monitored by an ISAAC data acquisition system which recorded all signals digitally. Finally, the ISAAC system was monitored by a micro-computer where signal conditioning took place. These signals were then passed via a modem to the mainframe computer. All EMG signals were recorded with recessed surface electrodes placed over the muscle of interest. The location of the electrodes was verified through functional muscle testing. The signal quality was also checked before and after each exertion.

Experimental task

The experimental task required the subject to be strapped to the reference frame with the knees in an erect position. The subject was required to bend forward to a 60 deg angle which was the starting point for all dynamic exertions. At this point the axis arm of the dynamometer was placed upon his back. Subjects were instructed to exert maximum extension force against the dynamometer throughout the range of trunk motion making sure they did not cease their effort until they had passed the 0 deg upright position. This task imposed loads upon the spine link which would be analogous to those experienced throughout a "back lift" motion.

Isometric exertions were measured in a similar manner. Subjects were placed in the experimental angle position with the dynamometer arm locked in place. They were then instructed to extend their trunk with maximum force against the dynamometer. Isometric exertions were three second in duration. All tasks require sagittally symmetric exertions.

All EMG data were normalized with respect to the maximum signal observed in the muscle of interest. Signal processing was accomplished on the micro-computer. The computer was also capable of analyzing the signal activity as a function of particular angle and velocity conditions (windows). All data were then placed in a large data base on the mainframe computer.

RESULTS

Muscle and torque response to motion

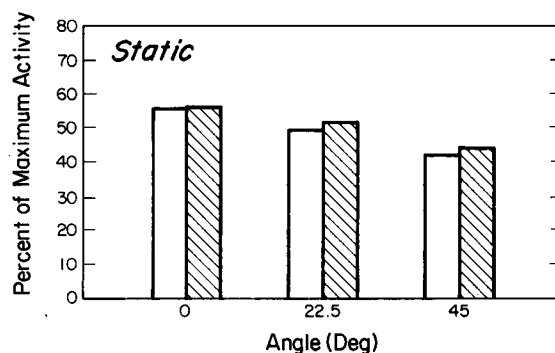The mean activity of the right and left latissimus dorsi muscle are shown in Fig. 1.
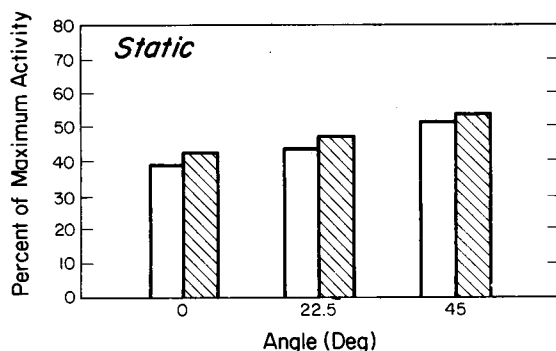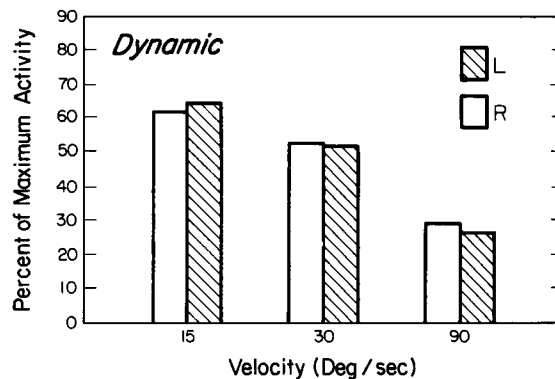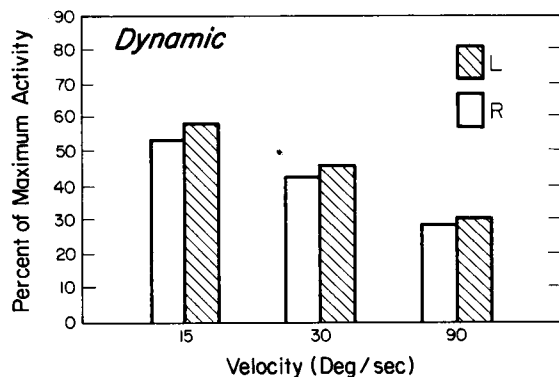
Fig. 1. Mean activity of the right and left latissimus dorsi muscle throughout each exertion.

Fig. 2. Mean activity of the right and left erector spinae muscles throughout each exertion.

This figure indicates the mean muscle force in each muscle throughout the duration of the exertion. Together, these muscles appear to increase their activity as the trunk angle increases under static conditions and decrease their activity as the trunk velocity increases under dynamic conditions. Under all trials the left muscle produced about three percent more activity than the right muscle in order to perform the sagitally symmetric task.

The mean activities of the right and left erector spinae muscles are displayed in Fig. 2. This figure also represents the mean force generation activity of each muscle throughout the duration of the exertion. Unlike the latissimus dorsi muscles this set of muscles decrease their activity as the static trunk angle conditions increase. These muscles also decrease their average activity under dynamic conditions as the rate of velocity increases.

The difference between activities on the right and left sides of the body were not as prominent (less than 2 percent) and did not exhibit a consistent pattern as did the latissimus dorsi muscles.

In both the latissimus dorsi and erector spinae muscles a unique pattern was noted when static trials were compared with dynamic trials. In both sets of muscles *the greatest amount of average muscle force produced throughout an exertion was exhibited under the 15 deg/s condition.*

The average absolute torque produced by the trunk throughout each experimental trial is shown in Fig. 3. This figure indicates that the average torque production capability was greatest when the trunk was in a static position of 22.5 deg or 45 deg. When the dynamic trials were considered, the torque production capability decreased as the velocity increased.
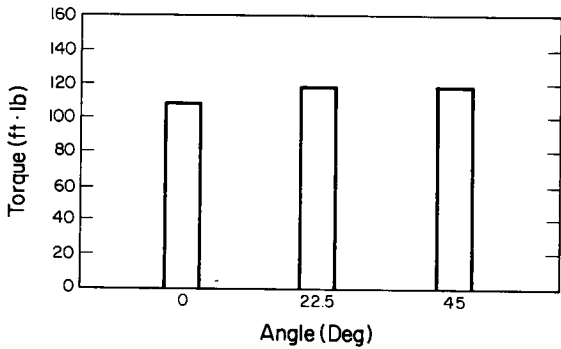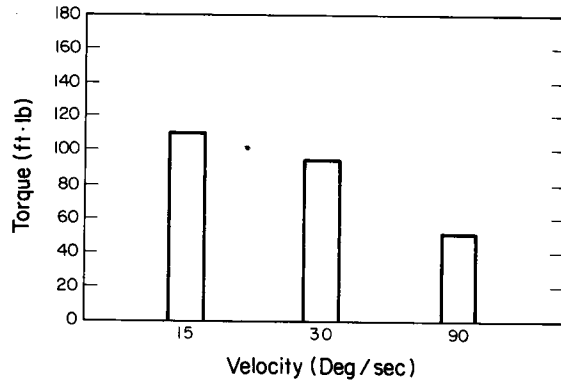
Fig. 3. Mean torque produced throughout each experimental trial.

The dependent variables behavior was also investigated as a function of the individual velocity–angle combinations as opposed to the trunk activity throughout the duration of

TABLE 1

ANOVA summary of significant responses to experimental conditions

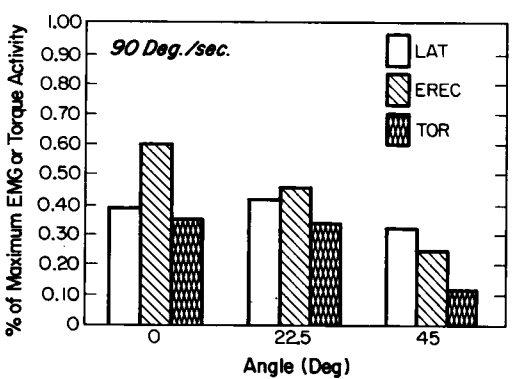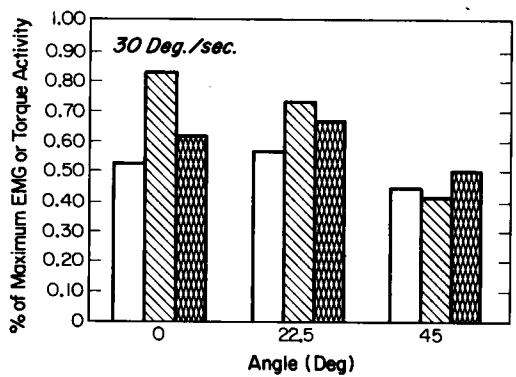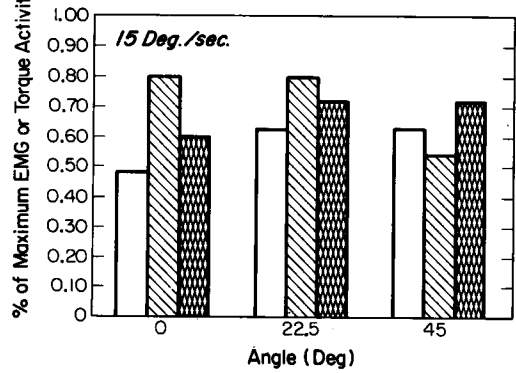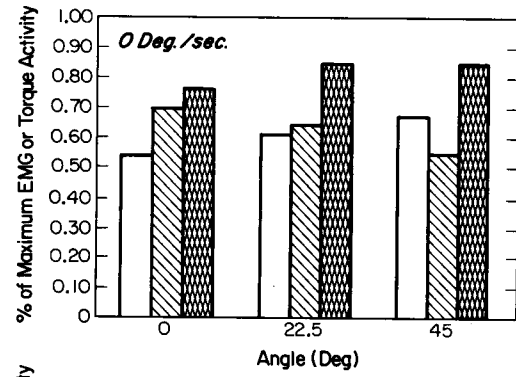| Variable	F value	Significance ($p < 0.01$)	R^2	C. V
Right Latissimus Dorsi	6.85	*	0.12	53.09
Left Latissimus Dorsi	9.11	*	0.16	48.79
Right Erector Spinae	40.09	*	0.45	29.48
Left Erector Spinae	51.03	*	0.51	28.32
Absolute Torque	41.78	*	0.465	33.70
Normalized Torque	86.34	*	0.64	27.76

Fig. 4. Internal and external loading as a function of the unique velocity and angle combination.

the exertion. The response of the dependent variables were observed at angle "windows" which were defined at the point where the experimental trunk angles occurred relative to the velocity condition: These responses were tested for statistical significance. Table 1 shows a summary of analysis of variance (ANOVA) tests performed on the dependent variables. All variables showed significant reactions to the experimental conditions. The erector spinae muscle and normalized torque were observed to account for a large amount of variability (R^2) in the experiment.

The average latissimus dorsi response, average erector spinae response and normalized (with respect to maximum) torque response as a function of the condition windows are shown in Fig. 4. This figure indicates that the erector spinae muscles produce greater activity at 0 and 22.5 deg angles under the slow velocity conditions (15 deg/s and 30 deg/s) than under the static conditions (0 deg/s). Torque production is also affected by velocity. Torque capability also drops substantially as a function of angle as velocity is introduced into the experimental condition. As shown in this figure, when even a small amount of velocity (15 deg/s) is introduced into the experimental condition a large amount (18–21%) of torque capability is lost at the various angles.

Relative motion costs to the muscles

The evaluation thus far has focused upon the response of individual dependent variables. However, it is evident that the various dependent measures each respond differently to increases in velocity. The dependent variables response averaged over all angles for the average latissimus dorsi signal, average erector spinae signal and normalized torque signal are shown relative to one another in Fig. 5. This figure indicates that average torque capacity decreases by approximately 0.55% of maximum for each deg/s increase in trunk
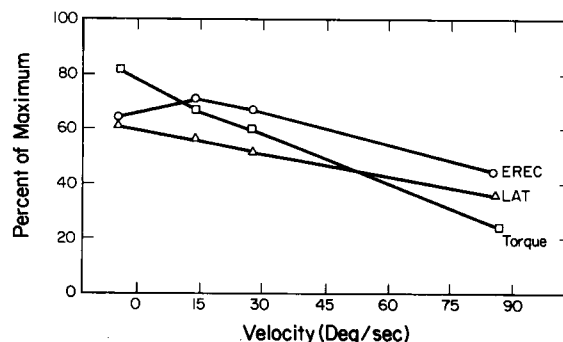

Fig. 5. Mean latissimus dorsi, erector spinae and torque activity averaged over experimental angles and shown as a function of velocity.

velocity. Latissimus dorsi activity decreases at a lesser rate than torque as velocity increases. However, the erector spinae activity increases as velocity increases at the slower velocities and then decreases at a slow rate when velocity increases. When these trends are considered collectively it is evident that dispropor-
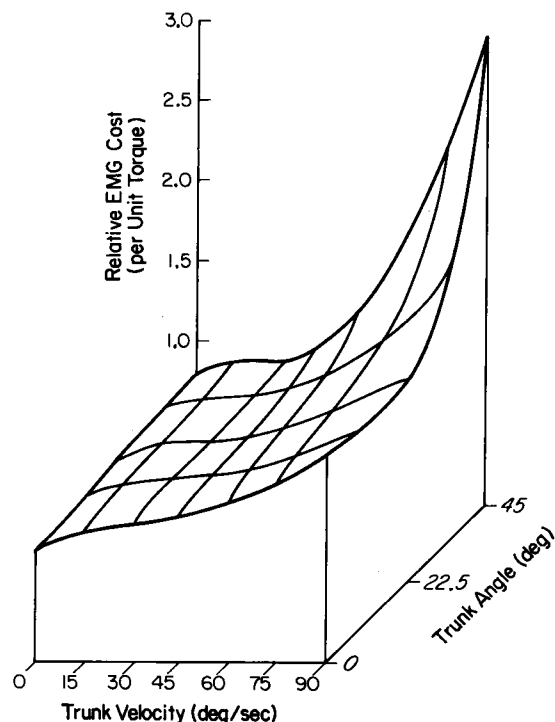

Fig. 6. Relative cost of motion and angle for the latissimus dorsi muscles to produce a constant amount of trunk torque.
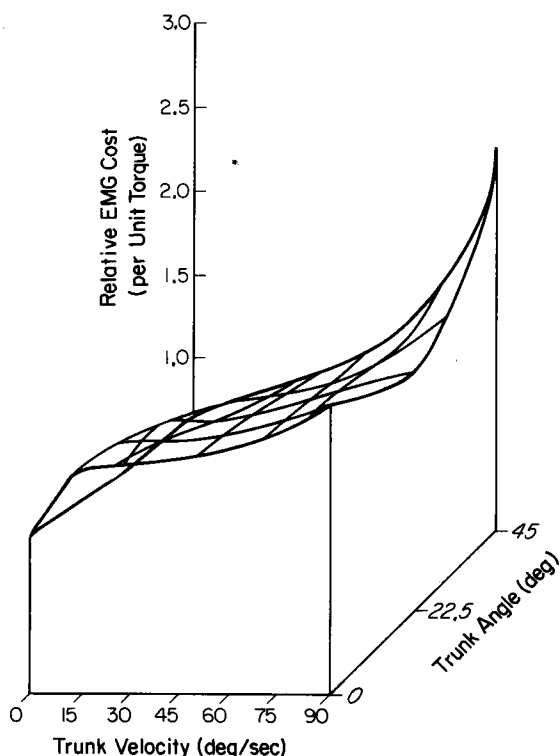
Fig. 7. Relative cost of motion and angle for the erector spinae muscles to produce a constant amount of trunk torque.

tional amounts of muscle force are required to produce a given amount of torque at the various velocities.

The response patterns described here indicate that the internal loading upon the spine due to the muscles' support of the trunk torque increases uniquely as the velocity conditions increase. One method to examine this factor is to observe the muscle force required to produce a standard amount of torque as a function of trunk angle and velocity. This relationship is shown for the latissimus dorsi signals and the erector spinae signals in Figs. 6 and 7 respectively. Each of these figures indicates that as the trunk velocity increases and the trunk angle increases, the cost to the muscle of producing a unit of torque increases dramatically. In fact when the trunk is moving at 90 deg/s the loading of the muscle

is well over twice that of the trunk in a static position.

Predictions of relative trunk loading costs

The transverse plane analysis technique of Schultz and Andersson (1981) was used to determine the relative influence of the back muscles in the internal loading of the spine. This method assumes that the cross-sectional area of the muscle and its relative distance from the center of the spine at a given vertebral level determines the contribution of the muscle to spine compression. Based upon computer tomography scans (CT scans) of the trunk at the third and forth lumbar vertebrae it was determined that the mix of erector spinae to latissimus dorsi muscle influence to total internal loading of the trunk was approximately a three to one ratio, respectively.

Since relative loading was of interest in this investigation the relative internal loading cost per unit torque, generated by the back was defined as a trunk loading variable. This was defined by calculating the loading forces experienced by the back (3 : 1 ratio of muscles) per unit torque. This variable represents the relative cost to the trunk of velocity and angle conditions.

Several regression analysis models were formulated to describe the back loading forces per unit torque experienced as changes in velocity and angle occurred. Table 2 shows a summary of significant response of these models to the experimental conditions. All models exhibited significant responses to the velocity and angle condition. However, the models which included all experimental conditions lack predictive power. They were able to explain less than 11% of the variability in the internal loading variable. These models also violated the normality assumption. The last two models in the table did not include the 90 deg/s condition. These models were capable of explaining approximately 50–60% of the variability in the data and do not

TABLE 2

Significance summary of model responses to velocity and angle for relative loading

Conditions included in model							Model structure	F value *= signif $p \geqslant 0.01$		Model R^2	Residuals normality
Velocity				Angle							
0	15	30	90	0	225	45					
×	×	×	×	×	×	×	linear	25.69	*	0.1044	0.7278
×	×	×	×	×	×	×	polynomial	13.35	*	0.1085	0.7299
×	×	×		×	×	×	linear	160.68	*	0.4934	0.9738
×	×	×		×	×	×	polynomial	116.57	*	0.5871	0.9663

violate the normality assumptions.

The regression model with the best predictive power was the polynomial model based upon all angle conditions and velocity conditions between 0 and 30 deg/s. A test of the regression components indicate that all components are significant except the A^2 component. Therefore, relative trunk loading cost can best be described by the following equation.

$$\text{Relative cost} = 0.405614 + 0.015477\ V - 0.00397\ A - 0.000436303\ V^2 \quad (1)$$

Equation (1) accounts for about 60% of subject variability between trunk angles of 0 and 45 deg and trunk velocities of between 0 and 30 deg/s.

DISCUSSION

This research has shown that significant changes in lifting capacity and the resultant relative loading of the spine occur when motion is considered in lifting tasks. These findings indicate that static evaluations of lifting capacity should be used with caution when attempting to evaluate a lifting task when even a small amount of trunk motion is involved. Based on this research, several adjustments should be made to evaluate the motion component in lifting.

First, a reduction in lift capacity was evident when motion was introduced into the experimental task. The reduction in torque generation capacity also depended upon the trunk angle. As the velocity increased the differences between torque generated at the experimental angles increased. Torque production at the 45 deg trunk angle decreased by the greatest amount as the velocity increased. This finding is particularly significant since the back force required during a dynamic lift is generally greatest at this point. This is due to the fact that under true dynamic lifting conditions the initial phase of the lift is involved in the production of acceleration necessary to overcome the inertia of rest of the load. Thus the required force is greatest at this angle, yet available torque production capacity of the trunk is relatively small.

The general torque loss experienced by the trunk as a function of velocity has been quantified. This value accounts for the average torque loss which can be expected over all angles. Generally, a reduction of 0.55% of maximum torque is experienced for every deg/s increase in trunk velocity. Marras and Wongsam (1986) have found that the mean velocity of the trunk was about 36 deg/s for normal healthy subjects lifting under leg lift and back lift conditions. Hence their actual maximum velocity would be about 80% of that predicted by static evaluations.

Next, this research has also shown that the cost of trunk loading due to the activity of the internal trunk supporting structures varies

greatly as the velocity of trunk motion increases. The cost per unit torque was defined as a measure of relative muscle force required to support a given constant load. The relationship between trunk velocity, trunk angle and the relative EMG force required to produce a standard unit of back torque was described in Figs. 6 and 7 for the latissimus dorsi and erector spinae muscles, respectively.

As an example of the biomechanical significance of the velocity effect, the spine loading due to the relative erector spinae force and load weight was calculated for the lifting situation shown in Fig. 8. This situation shows a subject lifting a 45 lb. weight which is at a horizontal distance of $1\frac{3}{4}$ ft. from the spine. This load is counterbalanced by the erector
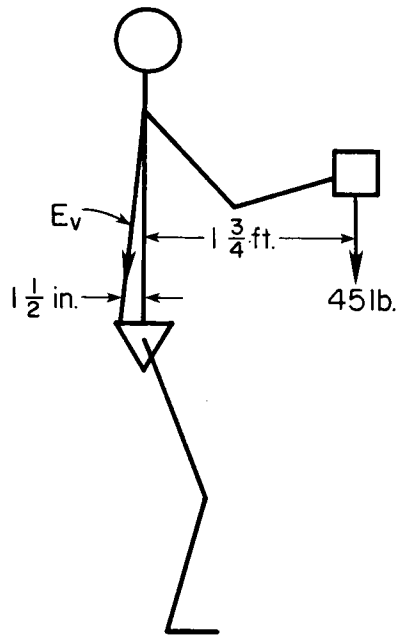
$$\tfrac{1}{8}\,\text{ft.} \bullet E_v \;=\; 1\tfrac{3}{4}\,\text{ft.} \bullet 45\,\text{lb.}$$

$$E_v = 630\,\text{lb.}$$

$$\text{Compression} = (E_v \bullet \text{Vel. Factor}) + \text{Load}$$

Fig. 8. A typical lifting analysis within the sagittal plane.

TABLE 3

Effects of velocity upon muscle force and L5/S1 compression computed as a function of the action of the erector spinae equivalent

Trunk velocity	EMG/ torque velocity factor	Muscle force equivalent E_V	L5/S1 compression
0 deg/s (static)	1.0 (reference)	630	675 lbs
15 deg/s	1.41	888.3	933.3
30 deg/s	1.43	900.9	945.9
90 deg/s	2.36	1486.8	1531.81

spinae muscle equivalent (NIOSH assumption) which has a 1.5-in.-moment arm from the spine. The minimum force required by the erector spinae muscles to support the load is 630 lbs. This force plus the load weight defines the compressive force on the spine. Realistically, the forces on the spine would be much greater than this prediction since the body segment weights and antagonists muscle actions have been ignored. Table 3 summarizes the results of this analysis. The velocity has increased the predicted compressive load upon the L5/S1 vertebrae. This compression has more than doubled over the 90 deg/s range. Comparing this analysis to the limits defined by the *Work Practices Guide for Manual Lifting* (NIOSH, 1981) emphasizes the importance of considering the velocity factor in work design. The static analysis produces a compressive load which is considered moderate by the NIOSH Guide. However, the introduction of even a small amount of velocity into the analysis increases the compressive loading by almost 40%. Including a relatively moderate amount of velocity increases the compressive load by more than a factor of two. The resultant compression would be considered dangerous according to the criteria set forth in the NIOSH Guide.

Finally, this research has also considered the relative back loading cost to the trunk velocity due to the combined influence of the

latissimus dorsi and erector spinae muscles. This relationship was described by the regression equation shown in eqn. (1). This equation was capable of describing a large amount of the subject variability and is capable of predicting the relative loading cost between 0 and 45 degrees of trunk angle at 0–30 deg/s of trunk motion. This equation could be used to predict the internal loading of the trunk which occurs due to the velocity and angle of the trunk throughout the lift. When this information is used in conjunction with body position and load information a better appreciation for the cost of lifting may be assessed. This equation may also be used to adjust the estimates of lifting cost predicted by NIOSH (1981). This could be accomplished in a manner similar to that shown in Fig. 8. The velocity of the trunk of workers could be assessed while performing a lifting task and the relative trunk loading would be predicted by including the trunk loading factor into the estimated back compression predicted by NIOSH.

The regression equation is valid for trunk velocities of 30 deg/s or less. Marras and Wongsam (1986) have shown that mean lifting velocities of workers assuming back and leg lifting postures without actually lifting a load were about 36 deg/s. This rate of trunk motion would be expected to decrease to within the range of eqn. (1) if measures were taken under actual load lifting conditions.

Future research is required to determine the components of asymmetrical motions. Also future research should evaluate the effects of these velocities in a manner similar to that described here for asymmetric motion ranges. In this manner more accurate assessments of the cost of dynamic lifting could be performed.

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