STOCHASTIC MODEL OF TRUNK MUSCULATURE DURING LIFTING

FINAL REPORT
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Project Summary

The broadly defined goal of this project was to explore the stochastic nature of the biomechanics of the low back during occupational manual materials handling tasks. Towards this end, three separate research projects were completed during the period of this grant. Project #1 was the specific study outlined in the original proposal. However, as is described below, Projects #2 and #3 became necessary as the model was refined. Project #4 describes a project just underway that is an application of some of the knowledge gained during the execution of this funded project. This final progress report has been divided into four sections - one for each of the research projects.

Project #1: Stochastic Model of Trunk Muscle Coactivation

Abstract

The specific aim of this project was to develop a stochastic model of the lumbar spine during lifting. The model developed in this research employs stochastic principles to predict the activation levels of ten trunk muscles under occupational lifting conditions including varied weights, postures, dynamic components and asymmetric lifting. Electromyographic (EMG) data were collected from 28 subjects as they performed simulated lifting tasks. These EMG data were collected from the right and left pairs of the erector spinae, latissimus dorsi, rectus abdominis, external obliques and internal obliques as subjects performed a variety of trunk extension exertions. These exertions were defined by different levels of trunk dynamics (isometric, isokinetic (10 or 45 deg/sec), constant acceleration (50 deg/sec/sec)), two levels of extension moment (30 Nm or 80 Nm), two levels of sagittal angle (5 and 40 degrees of forward sagittal bend) and two levels of trunk asymmetry (0 degrees and 30 degrees twisted to the right). Nine repetitions of each combination of independent variables were performed by each subject. The data collected during these trials were used to develop distributions of trunk muscle activity as well as a 10 x 10 correlation matrix that described how the muscles cooperated in the development of these extension moments. These elements were then combined to generate multivariate distributions describing the coactivation of the trunk musculature. The output of this project is an "EMG engine" that can produce muscle activities that would occur during occupational lifting.

Significant Findings

The product of this research project is a simulation model capable of producing simulated EMG traces given an input of trunk extension moment, instantaneous sagittal position, trunk extension velocity, trunk extension acceleration, asymmetry of posture. When this simulation model is run repeatedly it generates a range of activity levels for each of the primary trunk muscles. Figure 1 shows an example of the type of output that would result from repeated runs of the simulation model for the left erector spinae muscle These muscle activity levels can then be used to predict the spine reaction forces (compression and shear) that could occur during that particular lifting task.

Because this is a stochastic model this allows for a comparison of mean values with more extreme values. For example, the activity of the erector spinae was found to be, on average, 20-25% higher at a point in the distribution of 2 standard deviations above the mean as compared with the mean activity level. Since the erector spinae is the primary loader of the spine in compression this means that the compression on the spine can be as much as 25% greater than mean compression on 1 out of every 50 lifts. This type of variability has direct impact on musculoskeletal injury risk. While this level of muscular force may only occur infrequently, when it does occur it places a high load on the musculoskeletal system.

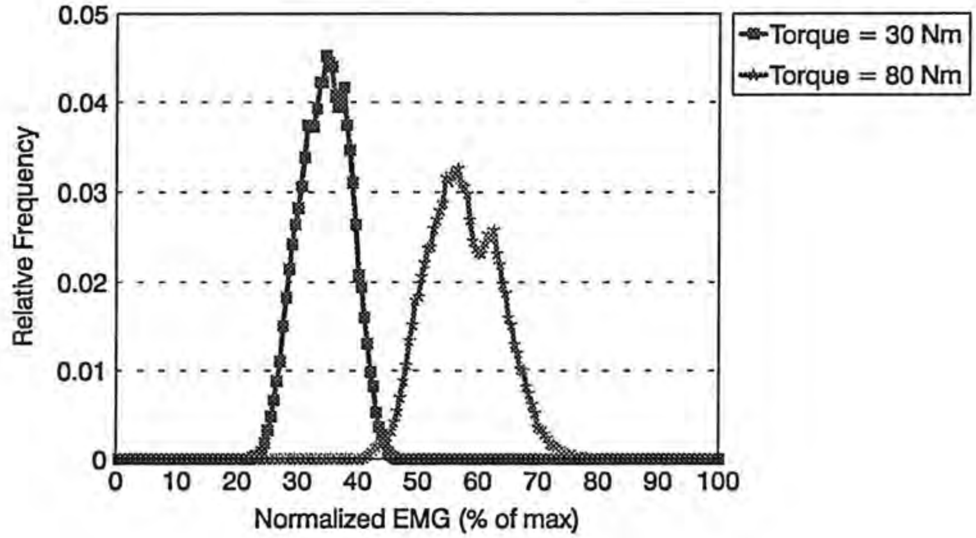

Figure 1. Distribution of Left Erector Spinae Muscle Activity: Sagitally Symmetric Lift, 10 deg/sec

Usefulness of Findings

Repetitive manual materials handling exists in many jobs in industry. Current modeling approaches would say that if the static configuration of the workplace (weight of load, origin of lift, destination of lift, frequency of lift) remains constant than the loading on the spine would remain constant as well. The results of this research disagree with this notion because they show that for a highly controlled repetitive lifting task there is a certain degree of variability in the way that the trunk muscles coactivate, which results in variability in the loading on the spine. Therefore, a basic usefulness of this work is simply to illustrate this stochastic dimension of manual materials handling that hadn't been previously explored or quantified.

A second, more practical use of this research is that the EMG engine that has been developed can be used to generate EMG data where EMG data collection is not feasible. Over the last decade a number of EMG-driven biomechanical models of spinal loading have been developed. The input requirements for these models are the activity levels of the primary trunk muscles. Most of this work has been limited to the laboratory because of the logistical and methodological constraints that an industrial environment presents in the gathering of EMG data. The EMG engine model approach could overcome some of these industry-based limitations and produce these values for input into the biomechanical models. All that is needed is a time dependent characterization of the moments and kinematics of the trunk and it will output estimates of the muscles' activity levels. These can then be input into the biomechanical models and estimates of the spine reaction forces can be established on the shop floor. This EMG engine is being utilized at North Carolina State University and at The Ohio State University.

Projects #2 and #3: Quantification of Variability in Trunk Kinematics during Manual Materials Handling Tasks

Abstract

As part of the original proposal it was suggested that the stochastic EMG engine model could be validated on the shop floor. As the research team began trying to implement the resulting model in industrial environments it became clear that in addition to the variability in the internal muscle forces, it was important to gain an understanding of the variability in the trunk kinematics during occupational lifting tasks. This resulted in two studies (Project #2 and Project #3) that quantified trunk dynamics during repetitive occupational lifting tasks. The specific aim of the first of these two studies was to quantify variability in the parameters describing sagittal plane motion during sagittally symmetric lifting tasks. The specific aim of the second of these two studies was to quantify the variability in 3-dimensional trunk kinematics during more realistic 3-dimensional lifting tasks.

In the first study, 7 male subjects performed simple sagittally symmetric lifting tasks. Subjects were asked to lift a weighted box at a rate of 4 lifts per minute for three minutes. The lifting tasks in this experiment were defined by combinations of three levels of coupling (poor, fair and good) and seven levels of load (4.5, 9, 13.5, 18, 22.5, 27 and 31.5 kg). As the subjects performed these lifting activities, the angular position, angular velocity and angular acceleration of the lumbar spine were monitored in the three cardinal planes of the body using the Lumbar Motion Monitor. These postural and dynamic measures were then used to calculate the peak moment about the lumbosacral joint in the sagittal plane. From these data the mean and variance of the parameters describing the trunk kinematics and trunk kinetics could be generated and an analysis of the effects of the task parameters on these measures was performed.

In the second study, 10 subjects (5 men and 5 women) performed more complex 3-dimensional lifting tasks. Subjects were asked to lift a weighted box (11.4 kg for men and 7.4 kg for women) from one conveyor to another, a motion that required a 90 degree turn. The lifting tasks in this experiment were defined by combinations of three levels of frequency (3, 6 and 9 lifts/minute) and two levels of starting height (30 and 60 cm). As the subjects performed these lifting activities, the angular position, angular velocity and angular acceleration of the lumbar spine were monitored in the three cardinal planes of the body using the Lumbar Motion Monitor. Each of these lifting tasks was performed for a period of twenty minutes. In addition to looking at the effects of the task parameters on the mean and variance of the parameters describing trunk kinematics, this data set also allowed us to see if time-dependent issues such as warm-up effects or fatigue effects might have an influence on either the mean or variance of these kinematic parameters.

Significant Results

Project #2

The results of this study showed that during highly controlled sagittally symmetric lifting increasing the load from 4.5 kg to 31.5 kg decreased the peak sagittal acceleration by 30% however total moment increased by 45% due to the increased weight held in hands. Good coupling tended to increase the peak sagittal acceleration but was impacted by an interaction with the load. The range of increase was between 5% (at 4.5 kg) up to 30% (at 22.5 kg).

The results of this study showed that during highly controlled sagittally symmetric lifting tasks there was significant variability in the magnitude of the peak velocity and acceleration in the sagittal plane. The results also showed that this variability was affected by the load held in the hands with the higher loads having greater variability (coefficient of variation for sagittal acceleration data increasing from 7 % at 4.5 kg up to 12 % at 31.5 kg).

Project #3

The results of the kinematic study of the 3-dimensional lifting tasks showed that both frequency of lift and starting height of lift had a significant effect on the peak sagittal acceleration. Reducing the starting height of the lift from 60 cm to a starting height of 30 cm produced a 120% increase in the sagittal acceleration and increased the peak sagittal angle from 23 to 55 degrees. Increasing the frequency of the lifts from 3 to 6 lifts per minute increased the peak sagittal acceleration by 7% but had no appreciable effect on the peak sagittal angle.

The time dependent traces of both the mean and standard deviation of sagittal acceleration showed subject-dependent trends over time. Illustrated in Figure 2 is the time dependent peak sagittal acceleration for one subject during one 20 minute bout of lifting. Figure 3 shows how the variability of the peak sagittal acceleration changed over time for one subject during one 20 minute bout of lifting.

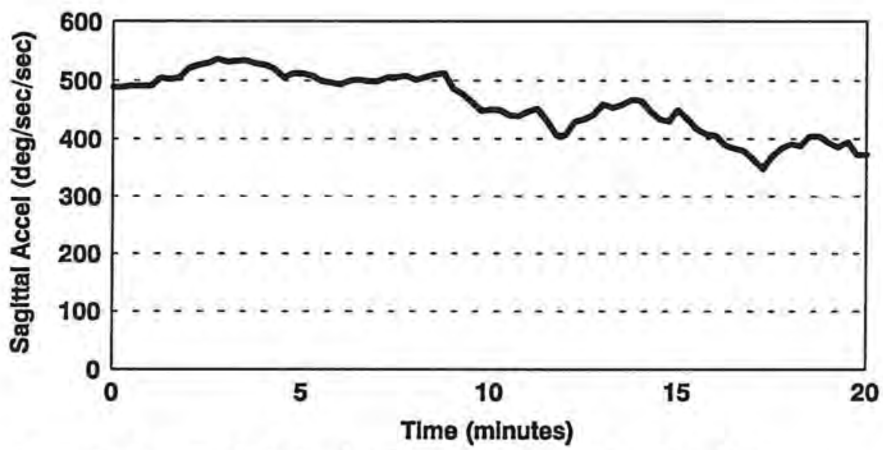

Figure 2. PeakSagittal Acceleration of Torso as a Function of Time Lift Frequency: 9 lifts/min, Starting Height: 30 cm.

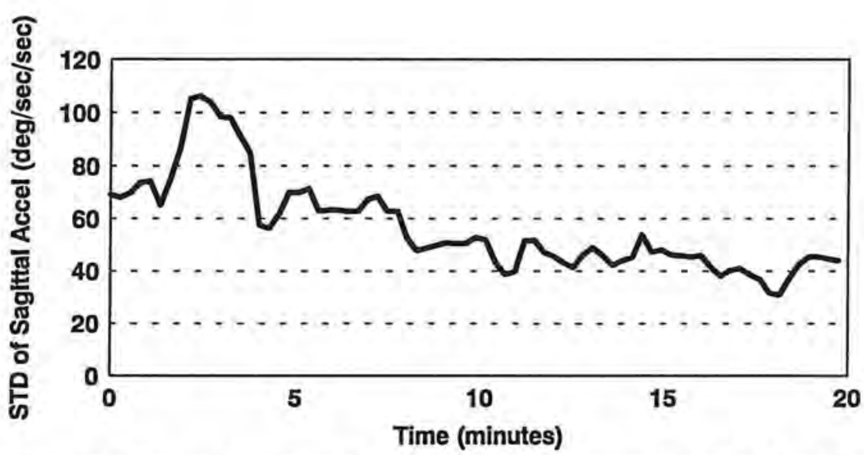

Figure 3. Standard Deviation of the Peak Sagittal Acceleration as a Function of Time. Lift Frequency: 6 lifts/min, Starting Height 60 cm.

Usefulness of Findings

One of the important things to consider when interpreting the results of this work is that the force on the spine is influenced by the acceleration of the torso and the peak sagittal angle of the torso. As a person performs a lifting task they are lifting not only the weight held in the hands but also the mass of their torso (estimated to be about 55% of the mass of the whole body). Since force is equal to mass times acceleration, it is important to try and minimize the acceleration of the torso in the sagittal plane to keep the muscle forces low which in turn will keep the spine compression low. Also of importance is the peak sagittal position of the torso, that is how far the lifter has to bend over. This is important because the moment arm of the center of mass of the torso is going to be a function of the peak sagittal angle. Therefore, our goal as ergonomists would be to keep the peak sagittal angle and the peak sagittal acceleration to a minimum.

Repetitive manual materials handling exists in many jobs in industry. Most of the current models used to asses risk of low back injury in the workplace do not consider the dynamics of the lifting motion when evaluating risk. Understanding the dynamics of the lifter and the variability in these dynamics may play an important role in understanding the risk of low back injuries. Therefore, a basic usefulness of this work is simply to illustrate this stochastic dimension of manual materials handling that hadn't been previously explored or quantified.

The data collected in this study can be used by researchers to estimate the kinematics of the torso during lifting tasks based on the static description of the workplace. As dynamic biomechanical models become more prevalent in ergonomic analysis, data describing trunk kinematics is going to become very important. One approach to this problem is to quantify the trunk kinematics directly using any of the commercially available motion analysis systems. However, these systems can be quite expensive. One of the practical applications of these studies is that they begin to develop a database of kinematic information describing the trunk motions that occur as an individual performs a lift, given a description of the task being performed (weight being lifted, starting point of lift, ending point of lift, frequency of lifting). This type of information will become useful as ergonomic analysis tools begin to incorporate biodynamics into their analysis.

Project #4: Development of a Methodology to Quantify Biomechanical Stress in Jobs with Variable Job Demands

Abstract

Extending the concept developed in Projects #2 and #3 to a more basic level, the specific aim of this project was to develop a modeling/assessment tool that will allow for a quantification of the biomechanical stresses that occur in jobs that have variability in their biomechanical demands. Whereas the previous projects looked at the variability in the performance of a simple repetitive job, this project recognizes that there are many jobs that aren't repetitive in this way and that many of the tools available for assessing biomechanical stress are not designed to address this issue of variable task demands. What is presented in this final report are the results of some pilot work which were used to support a proposal that was funded and is currently underway. The industry which is the focus of this research is the construction industry- specifically, small, residential contractors.

This stochastic exposure assessment model was applied to several jobs in the construction industry: mason, mason's helper, drywall installer, mortar mixer and framers (carpentry work). This technique involves documenting the manual materials handling tasks that the subjects perform and then evaluating those tasks with known assessment tools. Figure 4 show the results of this analysis using the NIOSH Revised Lifting Equation as the assessment tool. Plotted on this figure are the number of lifts at the different levels of the Lifting Index. Note the degree of variability present in the demands of this job. The objective of this research is to use this stochastic model of biomechanical stress to test the effectiveness of the ergonomic interventions developed for the construction industry.

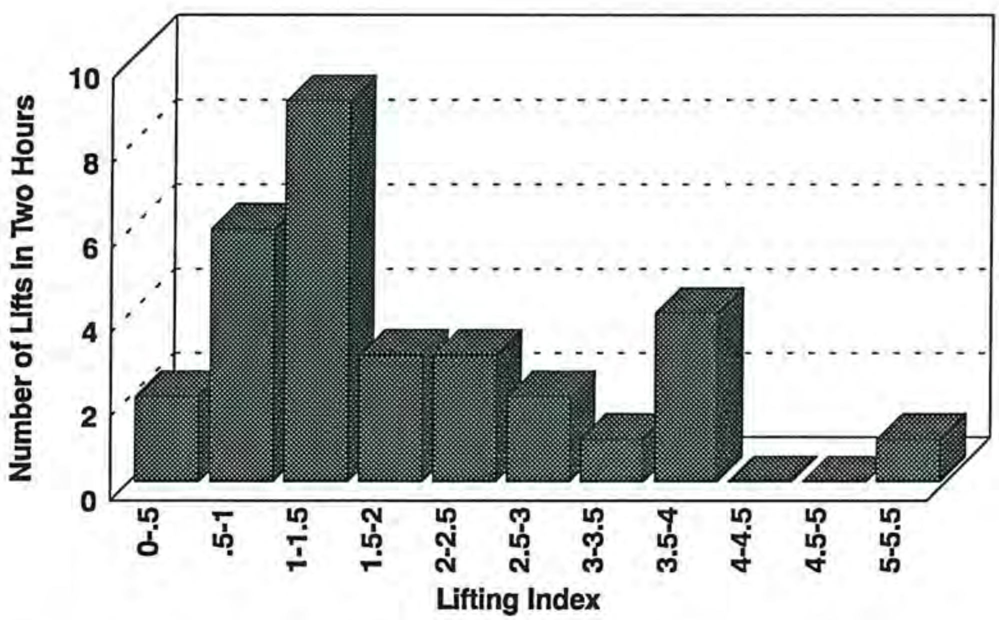

Figure 4. NIOSH Lifting Guide Assessment of a Sheet Rock Handler

Project #1

Publications

Mirka, Glasscock, Stanfield, Psihogios and Davis, "The Use of Multivariate Johnson Distributions to Model Trunk Muscle Coactivation", Proceedings of the Human Factors and Ergonomics Society 40th Annual Meeting, pp 584-588.

Mirka, Glasscock, Stanfield, and Psihogios, "The Use Of Multivariate Johnson Distributions To Model Trunk Muscle Coactivation", in Advances in Biomedical Engineering Vol 33 (S. Rastegar Ed.)) pp 299-300.

BACKGROUND

The incidence of occupation-related low back disorders (LBDs) has grown to epidemic proportions in the industrialized world. Epidemiologic studies indicate prevalence rates of between 12 and 35%, while lifetime incidence rates have been found to be between 50 and 70% (Andersson, 1981; Andersson et al, 1984; Biering-Sorenson, 1983; Frymoyer et al, 1983; Svensson and Andersson, 1982; Svensson and Andersson, 1983). Estimates of the cost of low back pain claims range from $6,807 to $30,000 per case and total cost to industry ranges from $11 to $30 billion annually (Industrial Commission of Ohio, 1987; Webster and Snook, 1990).

Many studies have indicated that the risk of LBD is associated with occupational factors; specifically, manual materials handling or lifting. The National Institute for Occupational Safety and Health (NIOSH, 1981) investigated the relationship between LBD and occupational lifting factors and found 60% of LBD claims were associated with overexertion. Further, there has been abundant research which has suggested a link between risk of injury during heavy work and the biomechanical stresses placed on the spine (Chaffin, 1969; Chaffin and Baker, 1972; Herrin et al, 1986; NIOSH, 1981).

In an effort to improve our understanding of occupation-related LBDs, biomechanical models of the lumbar region have been developed. Most models assume that external moments imposed about the spine are countered by the activity of the trunk musculature. Because of their mechanical disadvantage relative to external loads, the muscle forces are many times greater than the external loads and therefore become the primary loaders of the spine. The resultant vector created by the summation of these muscle forces must be resisted by the spine, resulting in compression, shear and torsional forces. Over time, the cumulative effect of these reaction forces is a gradual deterioration of the tissues of the spine. Hence, it is imperative to the understanding of spine loading (both acute and cumulative) that one accurately predict muscle behavior during the performance of occupational tasks.

Over the last 20 years, the development of these models has gone through a gradual metamorphosis. Early formulations consisted of single equivalent muscle models of the trunk extensors. Chaffin (1969) and Chaffin and Baker (1972) described the muscle forces necessary to counteract a load held in the hands during a sagittally symmetric isometric exertion. Their approach was to develop a simple two-dimensional free body diagram utilizing a single extensor muscle equivalent that supplied the restorative moment needed to counter the external load. They first calculated the muscle force of the single extensor muscle equivalent and then calculated the spine compression using simple mechanical principles. This spine compression value has traditionally been used as the primary spine reaction force to establish risk level of a job. For example, the Work Practices Guide for Manual Lifting (NIOSH, 1981) determined that one of the key elements in deciding the acceptability of a given task was the compression experienced by the spine. It was generally held that compressions below 3400 N were considered safe while those above 6400 N were considered hazardous. These limiting values for compression were derived from cadaver studies of tissue tolerances (Evans and Lissner, 1959; Sonada, 1962). It is interesting to note, however, that there have been no validation studies which have shown that these guidelines have had a significant impact on the incidence of low-back injury.

More recent models have developed the anatomy of the trunk musculature more fully to allow for multiple trunk muscles (Schultz and Andersson, 1981). In these three dimensional models, not only is there compression in the spine but now the models are capable of calculating the anterior/posterior (A/P) shear and the lateral shear forces which result from the summation of the three-dimensional muscle force vectors. A problem with these models is that because there are many unknown muscle forces and only six static equilibrium equations the system is statically indeterminate. This condition requires that the model make some assumptions about the relative contribution of the trunk muscles. Some of these models use optimization techniques such as linear programming while others simply assume that some of the muscles are inactive. These techniques, in general, have resulted in muscle force predictions which have been shown to be inaccurate, especially under non-steady state, dynamic conditions.

It is believed that the limitation of these biomechanical models of the spine is that they have assumed that the forces exerted by the muscles of the trunk during lifting motions are deterministic in nature. That is, given a set of lifting conditions (weight, posture, motion, etc) there is one preferred combination of muscle forces which will be employed by the body to generate the lifting moment. However, the indeterminate nature of the biomechanical system implies that each individual trunk muscle can be variable in the amount of muscle force it exerts on any given lifting exertion. It is hypothesized that this variability is a key component in understanding risk of injury to the spine because each trunk muscle has its own distinct line of action which is composed of compressive, A/P shear and lateral shear components. Thus, if on one trial a muscle with a high lateral shearing component is inactive while on another trial it is active, the variability of this muscle's activity has a direct bearing on the loading pattern of the spine, which is of prime importance when assessing risk of manual materials handling jobs.

Two studies (Herrin et al, 1986; Marras et al, 1993) have shown through biomechanical analyses of the work place that risk of LBD was more a function of the peak, yet often infrequent, loads imposed upon the spine. This implies that it is essential to document not only the average spine loading associated with a task, but to understand the peak spine loading as well. This is where the muscle force variability becomes important. If an accurate estimation of muscle force variability can be developed, our ability to predict the magnitude of these extreme spine loads as well as the frequency at which they occur will enhance our understanding of the cumulative trauma to the spine.

A group of biomechanical models that have been shown to be more accurate in their predictions of muscle forces under dynamic conditions are the EMG-assisted biomechanical models. The basic premise of these models is that instead of estimating muscle forces, these models utilize an indirect measure of muscle force, specifically, electromyography (EMG). These models use these time dependent muscle activity values to establish a biomechanical balance in the trunk and then calculates the spine reaction forces which must result. Because they are data driven, these models also have the unique ability to model the variability of the biomechanical systems. However, in order to accomplish this goal, one would need to have a subject perform a lifting or bending motion repeatedly and then run the data through the biomechanical model. Further, this method has only limited application because for every task one wished to model, another experiment would have to be devised, subjects run and data analyzed. This is not a very efficient laboratory practice and would be impossible on at the work site because of practical considerations of collecting EMG data outside the laboratory.

A more practical approach is to develop a predictive model of muscle forces which uses stochastic principles in arriving at its predictions. This model would be able to predict muscle forces under a variety of realistic lifting conditions and could be used to drive any of the EMG-assisted biomechanical models which have been developed.

SPECIFIC AIMS

The specific aim of this research was to develop a simulation model of the lumbar trunk that utilizes stochastic principles in determining forces acting about and within the spine. In doing so, it is believed that a more realistic assessment of risk for low back disorders can be established for manual material handling tasks.

METHODOLOGY

Subjects

Twenty eight people from the university community served as subjects in this study. There were twenty one men and seven women. None of the subjects had a history of low back disorders (defined as no lost time from work or school due to back pain) and each signed an informed consent form before participating in this study. Experience in manual material handling tasks varied. Basic subject anthropometry is listed in Table 1.

Table 1. Basic Anthropometry of Subject Population

Variable	**Mean**	**Standard Deviation**
Age (years)	29.43	8.65
Body Mass (kg)	78.2	14.4
Height (cm)	175.4	8.9

Apparatus

A Kin/Com dynamometer was used in conjunction with a trunk motion reference frame to provide an environment that allowed the researchers to have a great deal of control of the extension moments, postures and movements of the subjects. (See Figure 1.) An EMG data processing system and a data collection system were used to gather the data describing the signals from the dynamometer (2 load cells, position potentiometer and velocity tachometer) and the muscle activity levels (ten trunk muscles). The EMG signals collected by the electrodes were amplified 1000x by miniature preamplifiers located at the muscle site. The electrode leads to the preamplifiers were kept short so as to reduce the movement noise and the external electrical noise from the surrounding environment. The signal was amplified (total amplification ~ 60,000x) and high and low pass filtered at 80 and 1000 Hz. This filtered signal was rectified and processed using a 20 msec moving average window. These processed EMG data along with moment, angle, and velocity were collected at 100 Hz by the data collection system.

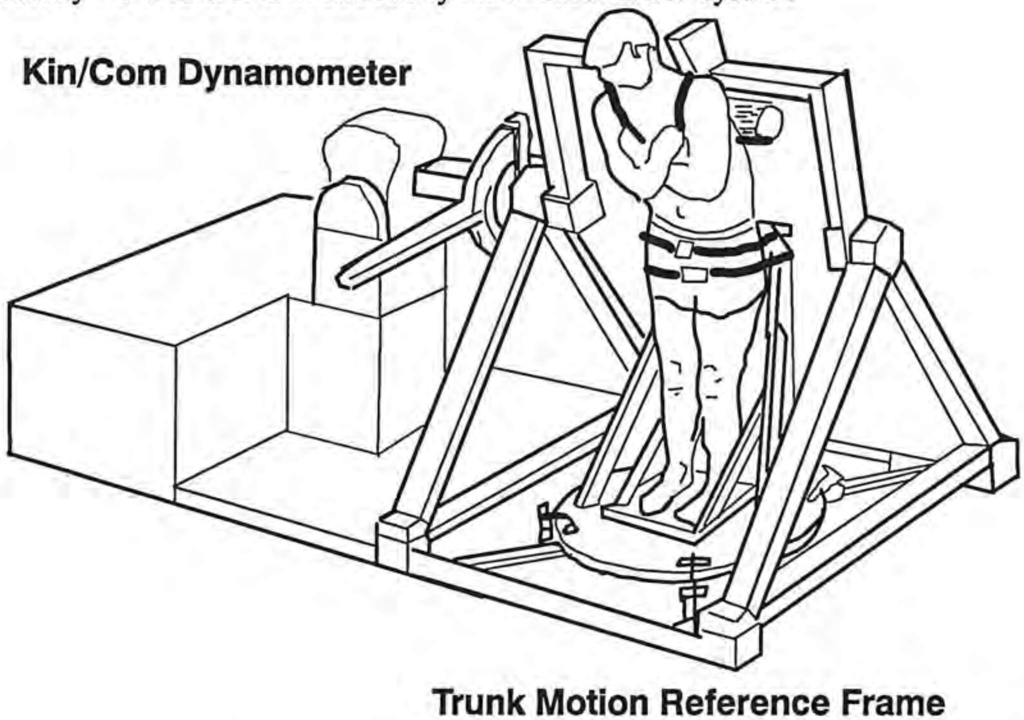

Figure 1. Experimental Apparatus (Trunk Motion Reference Frame and Kin/Com dynamometer)

Experimental Design

Independent Variables. In order to quantify the variability of the muscle forces during lifting, the subjects in this experiment were asked to perform highly controlled trunk extension motions repeatedly. These trials included isometric, isokinetic (10 or 45 deg/sec) and constant acceleration (50 deg/sec/sec) exertions. Moment exerted by the subjects were either 30 Nm or 80 Nm for the experimental trials. (For five of the subjects the 80 Nm condition was beyond their capability and therefore the upper force level was reduced to 60 Nm for those subjects.) Two trunk positions (5 and 40 degrees of forward sagittal bend) and two levels of trunk asymmetry (0 degrees and 30 degrees twisted to the right) were evaluated in this study. Each combination of independent variables was repeated 9 times per subject. The order of presentation of the combinations of the above independent variables was completely randomized within level of asymmetry and the presentation of asymmetry was counterbalanced across subjects.

Dependent Variables. The dependent variables in this study were the normalized processed EMG values of the ten trunk muscles identified by the transverse cutting plane technique described by Schultz and Andersson (1981). These muscles included the right and left erector spinae (RES, LES), right and left latissimus dorsi (RLAT, LLAT), right and left rectus abdominis (RAB, LAB), right and left external obliques (REX, LEX) and the right and left internal obliques (RIN, LIN) muscles. The inter-electrode distance for each electrode pair was 3.0 cm.

Procedure

Upon arrival the subjects had surface electrodes applied to their skin through standard preparation procedures (Marras, 1990). The subject was then asked to enter the reference frame so that the adjustable base could be set for the subject's leg length in order to insure that the subject's L5/S1 joint was aligned with the rotating axis of the Kin/Com dynamometer. Once the subject was secured in the reference frame they performed maximum voluntary contractions (MVCs) at four positions (5 and 40 degrees of sagittal bend and 0 and 30 degrees of asymmetry). Both maximum static extensions and flexions were collected as well as the resting values in each of these postures. After these maximal exertions, the experiment began with the subject performing a sequence of randomized trials. Each of these trials dictated that the subjects perform a controlled exertion defined by set levels of moment, posture, angular trunk velocity, and angular trunk acceleration. During these trials the angular position, velocity and acceleration were controlled by the dynamometer. The exerted moment was controlled by the subject within a tolerance of +/- 10% using a graphical video feedback system that displayed their instantaneous moment output as well as the target moment designated for the particular trial. If the subject failed to maintain the designated extension moment the trial was repeated.

Data Analysis

The EMG data were first normalized with respect to the maximum and resting EMG values that occurred at each particular trunk posture. The main emphasis of this research project was to better understand the effects of the task parameters on the distributions of muscle activity. We were therefore interested in eliminating the inter-subject variability. This was accomplished by standardizing the data across subjects so that the variability between subjects would not influence the results. This was accomplished by calculating a mean and a standard deviation for each subject in each experimental condition. The overall mean and pooled standard deviation were then calculated for each condition. Using these values, the individual EMG values were then standardized using the following formula:

$$SV(j, k, l, m) = MP(j, k) - [STDP(j, k) * (M(j, k, l) - AV(j, k, l, m))] / STD(j, k, l)$$

Where:

SV (j, k, l, m) - standardized EMG value of muscle j, condition k, subject l and repetition m
AV (j, k, l, m) - actual EMG value of muscle j, condition k, subject l and repetition m
STDP (j, k) - pooled standard deviation for muscle j and condition k
STD (j, k, l) - standard deviation for muscle j, condition k and subject l
MP (j, k) - average for muscle j and condition k
M (j, k, l) - average for muscle j, condition k and subject l

Model Development

At this point the data was in the form of 32 - {10 X ROW} matrices containing the normalized, standardized EMG values, where 10 refers to the 10 muscles sampled and ROW refers to the number of trials that met the strict criteria laid out for the acceptability of the data based on the position, velocity, acceleration and moment parameters for the trial. The range across experimental conditions for the number of acceptable trials was 102-180. The 32 different matrices refer to the 32 unique combinations of independent variables that each of the subjects performed.

Each of these 32 data sets were then used to generate a set of multivariate distributions. The procedure used is described in greater detail in Stanfield (1993) and is briefly outlined below.

1) Determine the first four moments of the distribution for each muscle (mean, standard deviation, skewness and kurtosis) and the correlation coefficients between muscles.

2) Develop a lower triangular matrix V such that $V V^T = C$, where C is the {10 X 10} correlation matrix.

3) Develop two new standardized {1 X 10} skewness and kurtosis vectors using the following equations:

   $s^* = (V^{(3)})^{-1} * s$ where s is the original {1 X 10} skewness vector
   $k^* = (V^{(4)})^{-1} * [k - 6 * \Sigma^{9}_{j=1} \; \Sigma^{10}_{l=j} V^2_{ij} * V^2_{il}]$ where k is the original {1 X 10} kurtosis vector

4) Using the above standardized skewness and kurtosis vectors, fit a marginal Johnson distribution to each of the muscle distributions (DeBrota et al, 1989).

5) Finally, to generate samples that reflect the true multivariate nature of the data use the following relationship:

   $$X = S (V * Y) + \mu$$

   Where:
   - X is a {1 X 10} vector of actual multivariate values
   - S is a {10 X 10} diagonal matrix containing the original standard deviation for each muscle
   - V is the {10 X 10} lower triangular matrix as described in 2) above
   - Y is a {1 X 10} vector of samples from the marginal distributions generated using the Johnson distributions developed in 4) above.
   - μ is a {1 X 10} vector of the original means

Using the above outlined procedure, multivariate Johnson values are generated for each muscle under each experimental condition. With multiple runs of the simulation, the shapes of the best fit distributions were developed.

RESULTS

The output from a sample run of the simulation model is shown below. The data presented is for the case when asymmetry = 0 degrees, sagittal angle = 40 degrees, sagittal velocity = 10 degrees/sec, sagittal acceleration = 0 (isokinetic), and extension moment = 30 Nm.

First, for each of the muscles the mean, standard deviation, skewness and kurtosis are found.

Muscle	Mean	Standard Dev	Skewness	Kurtosis
R. Rectus Abdominis	0.025991	0.008783	0.334692	2.472747
L. Rectus Abdominis	0.025699	0.008226	0.365884	2.690220
R. External Oblique	0.035872	0.013182	0.243744	2.476756
L. External Oblique	0.035788	0.009580	0.347826	2.399736
R. Internal Oblique	0.076432	0.028867	0.392143	2.638061
L. Internal Oblique	0.081094	0.025693	0.202989	2.194189
R. Latissimus Dorsi	0.131311	0.024074	-0.070839	2.354376
L. Latissimus Dorsi	0.155298	0.023341	-0.030379	2.415294
R. Erector Spinae	0.362318	0.042168	0.005781	2.204787
L. Erector Spinae	0.390794	0.049695	0.159614	2.349448

Next, the correlations between the muscle activities were found.

Correlation Matrix

1.	.5018	.4396	.3615	.1577	.1433	.0270	.0583	.0061	-.0195
.5018	1.	.4708	.4002	.1066	.1705	.0639	-.0177	.0185	.0744
.4396	.4708	1.	.5648	.106	.1926	.1024	.1122	-.0289	.0808
.3615	.4002	.5648	1.	.1647	.2786	.1494	.1776	.0726	.0540
.1577	.1066	.106	.1647	1.	.484	.4252	.1254	.1022	.0856
.1433	.1705	.1926	.2786	.4840	1.	.3607	.3915	.1245	.1406
.0270	.0639	.1024	.1494	.4252	.3607	1.	.4580	.4957	.3173
.0583	-.0177	.1122	.1776	.1254	.3915	.4580	1.	.3263	.4496
.0061	.0185	-.0289	.0726	.1022	.1245	.4957	.3263	1.	.5405
-.0195	.0744	.0808	.0540	.0856	.1406	.3173	.4496	.5405	1.

This correlation matrix was used to find the V matrix as described in 2) above.

V Matrix

1.	.	.	.	.	.	.	.	.	.
.5018	.8650	.	.	.	.	.	.	.	.
.4396	.2893	.8503	.	.	.	.	.	.	.
.3615	.2530	.3913	.8076	.	.	.	.	.	.
.1577	.0317	.0324	.1077	.9805	.	.	.	.	.
.1433	.1140	.1136	.1901	.4422	.8496	.	.	.	.
.0270	.0582	.0867	.1126	.4122	.1609	.8831	.	.	.
.0583	-.0543	.1203	.1525	.0995	.3562	.3778	.8224	.	.
.0061	.0178	-.0432	.1025	.0928	.0777	.4936	.1132	.8463	.
-.0195	.0973	.0720	.0103	.0838	.1001	.2877	.3563	.4053	.7706

Finally, the distributions are fit to the Johnson distributions which are described by four parameters: Gamma, Delta, Psi and Lambda.

Muscle	Gamma	Delta	Psi	Lambda
R. Rectus Abdominis	0.587	1.227	-2.304	5.804
L. Rectus Abdominis	0.700	1.006	-1.848	5.141
R. External Oblique	0.353	0.738	-1.653	4.010
L. External Oblique	0.388	0.248	-0.946	2.619
R. Internal Oblique	0.775	1.296	-2.296	6.193
L. Internal Oblique	0.190	0.329	-1.189	2.736
R. Latissimus Dorsi	-0.164	0.768	-2.182	4.041
L. Latissimus Dorsi	-0.067	0.663	-1.908	3.685
R. Erector Spinae	0.030	0.453	-1.496	3.051
L. Erector Spinae	0.235	0.201	-1.004	2.434

These are then used to generate samples that reflect the true multivariate nature of the data use the following relationship as described in 5) above.

The results of this simulation are distributions for each of the trunk muscles in each of the experimental conditions. Displayed in Figures 2 - 5 are a small sample of these fitted distributions. Figures 2 and 3 show the best fit distributions for the right erector spinae and the right rectus abdominis muscles while Figures 4 and 5 show the best fit distributions for the right latissimus dorsi and the right external oblique muscles, respectively. Note how this distribution fitting system models even the very skewed distribution of the right latissimus dorsi.

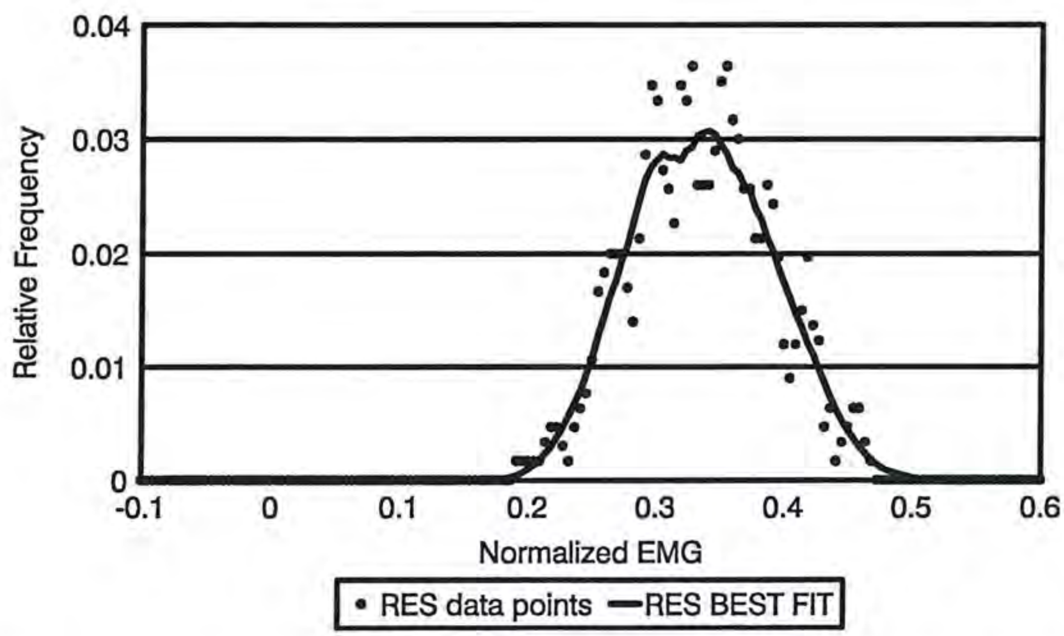

Figure 2. Empirical data and best fit distribution for the right erector spinae (Sagittal Angle = 40, Sagittal Velocity = 10 deg/sec (isokinetic), Extension Moment = 80 Nm)

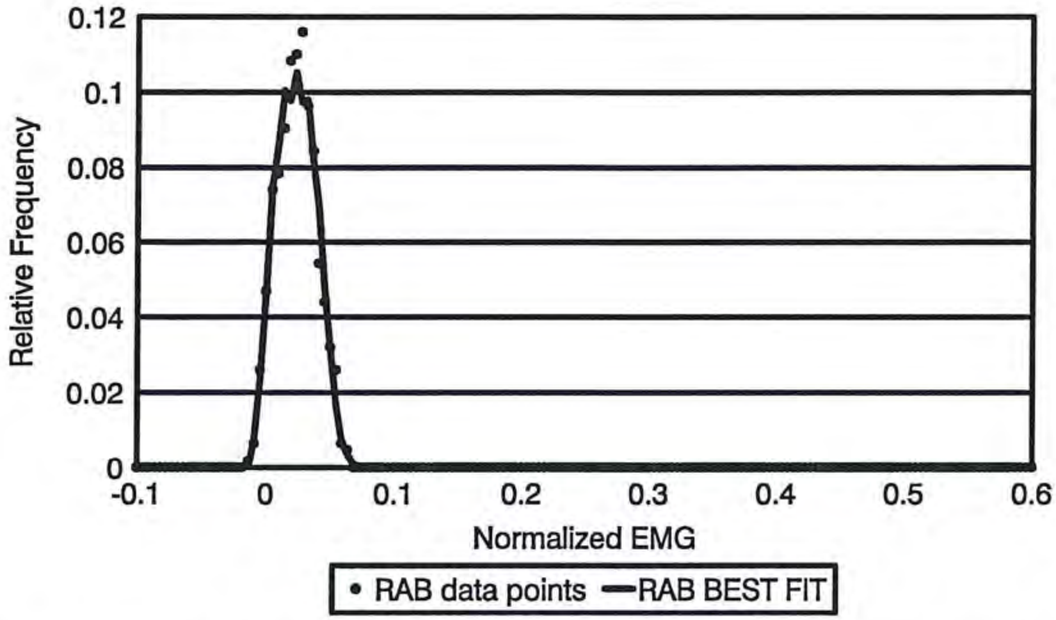

Figure 3. Empirical data and best fit distribution for the right rectus abdominis (Sagittal Angle = 40, Sagittal Velocity = 10 deg/sec (isokinetic), Extension Moment = 80 Nm)

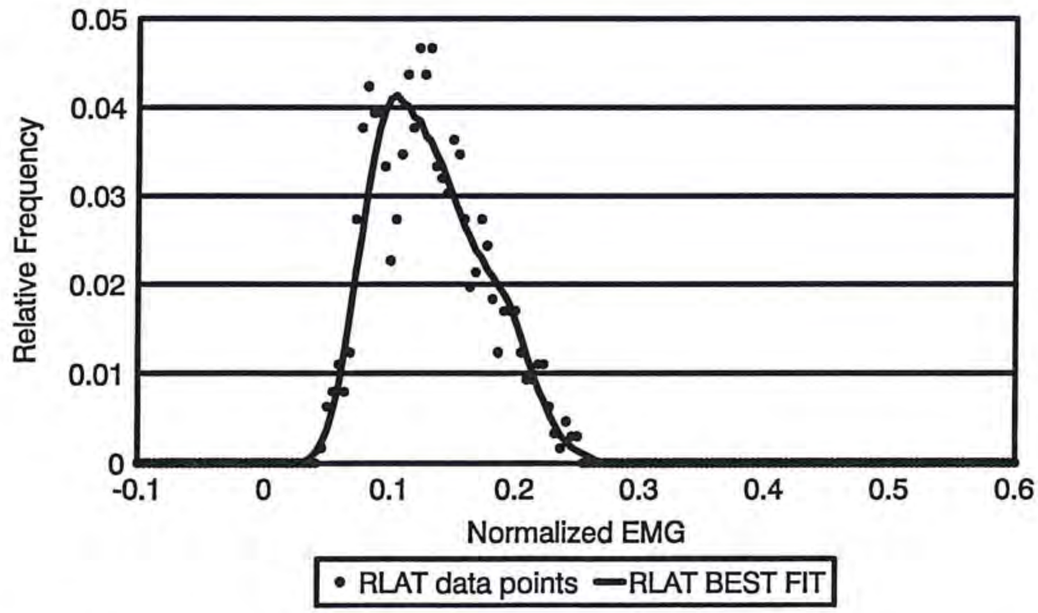

Figure 4. Empirical data and best fit distribution for the right latissimus dorsi (Sagittal Angle = 40, Sagittal Velocity = 10 deg/sec (isokinetic), Extension Moment = 80 Nm)

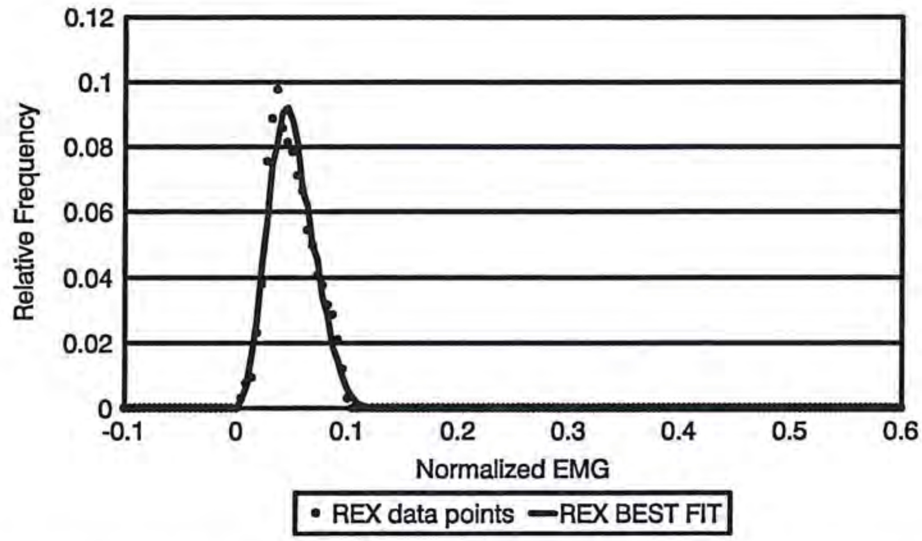

Figure 5. Empirical data and best fit distribution for the right external oblique (Sagittal Angle = 40, Sagittal Velocity = 10 deg/sec (isokinetic), Extension Moment = 80 Nm)

To gain an appreciation for the amount of variability present in the data, sets of input parameters (extension moment, angle of asymmetry, sagittal position, sagittal velocity and sagittal acceleration) were input into the simulation model and then the simulation model was run 5000 times. The results describe, for a given lifting task, the amount of variability that may exist in the muscular activations as a function of workplace parameters. A sample of this output is presented here in Figures 6-11.

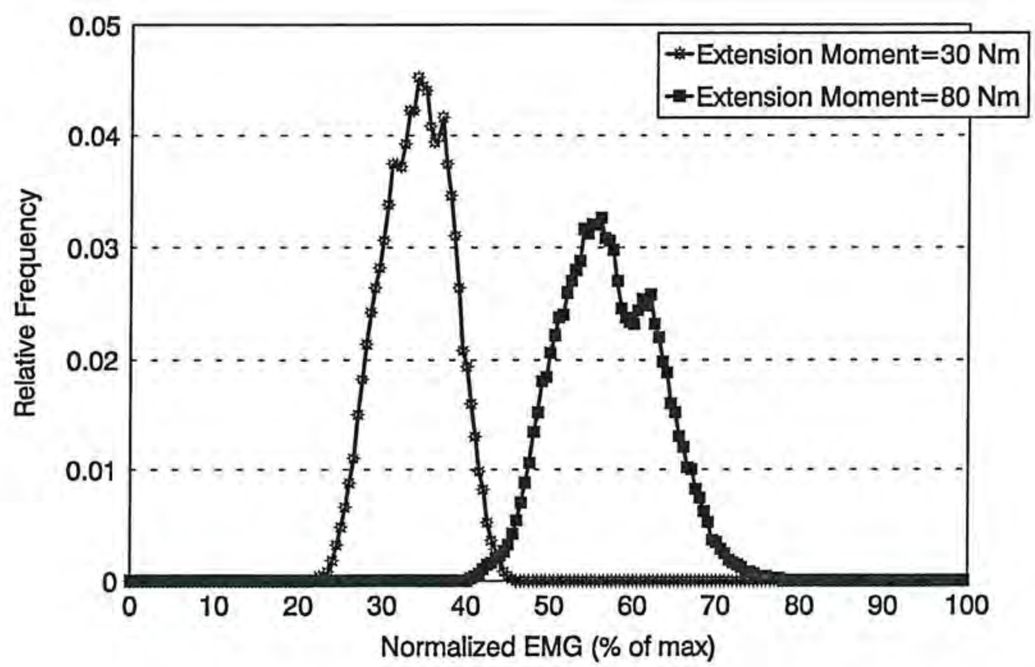

Figure 6. Distributions of the Left Erector Spinae as a function of extension moment. (Sagittal Angle = 5 deg, Sagittal Velocity = 10 deg/sec (isokinetic), Sagittally Symmetric)

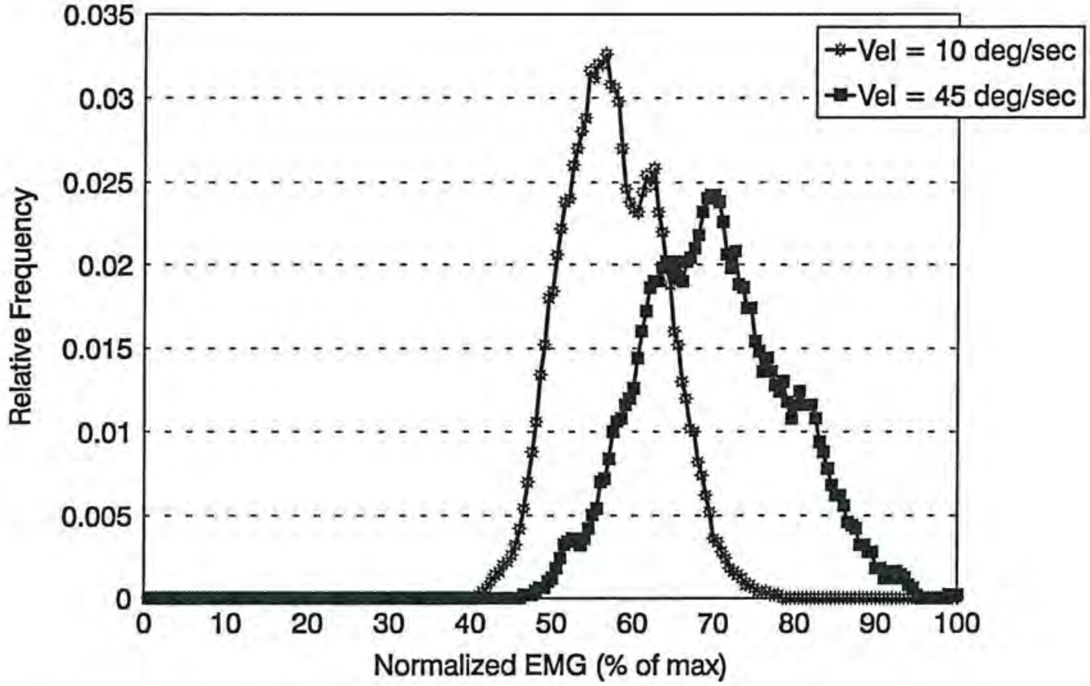

Figure 7. Distributions of the Left Erector Spinae as a function of extension velocity. (Sagittal Angle = 5 deg, Isokinetic, Extension Moment = 80 Nm, Sagittally Symmetric)

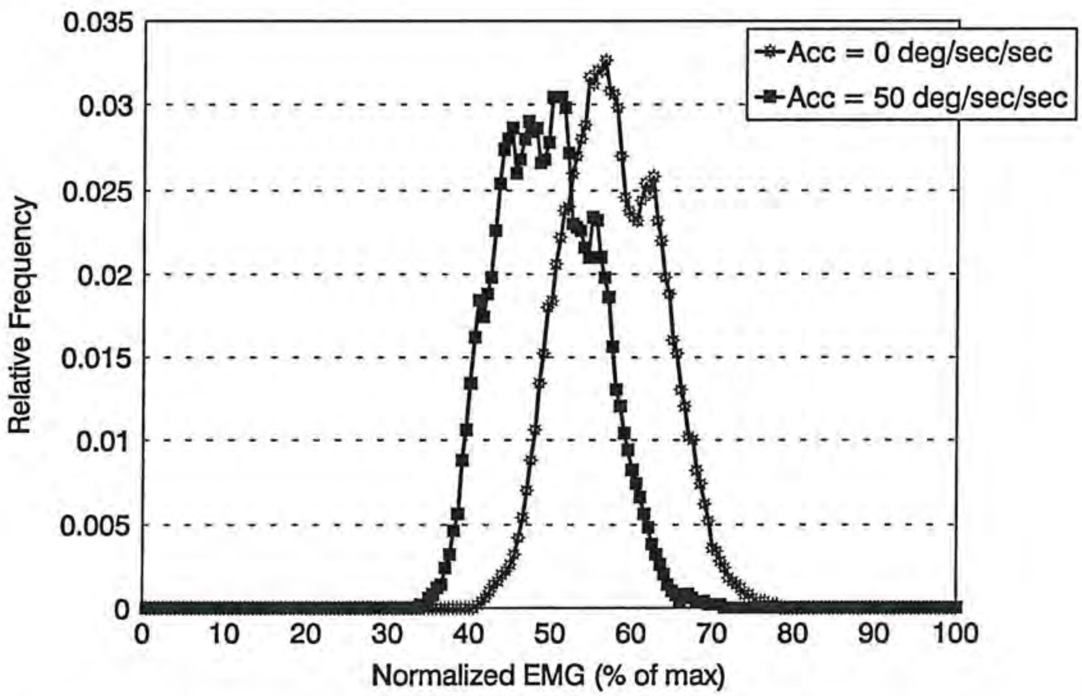

Figure 8. Distributions of the left erector spinae as a function of sagittal acceleration. (Sagittal Angle = 5, Extension Moment = 80 Nm, Sagittally Symmetric)

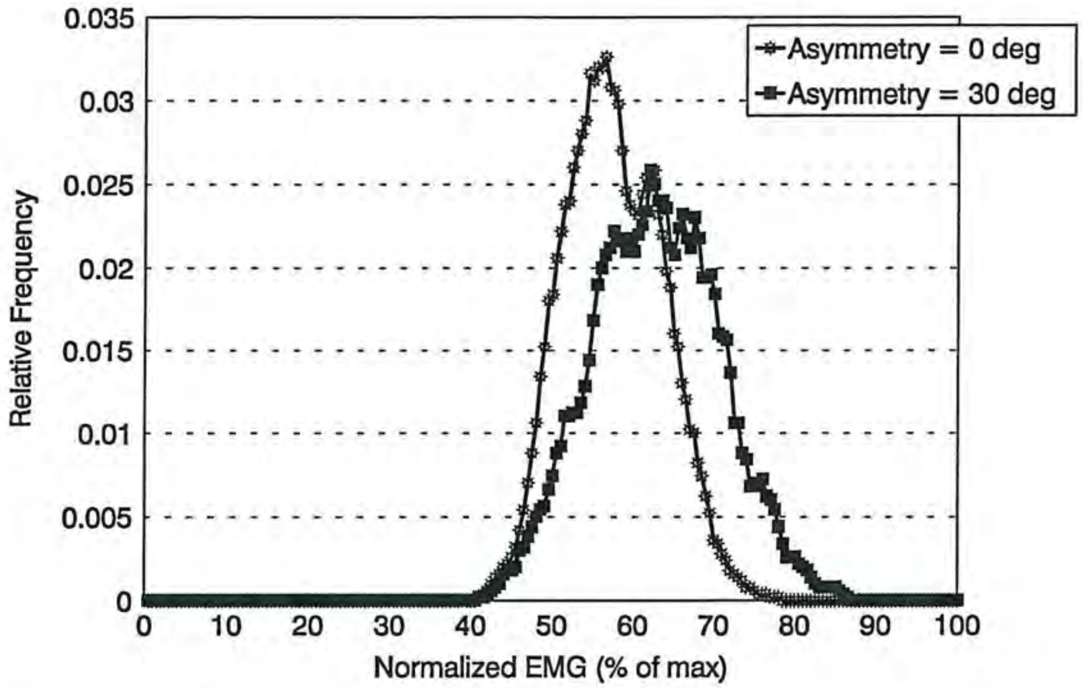

Figure 9. Distributions of the left erector spinae as a function of asymmetry. (Sagittal Angle = 5, Extension Moment = 80 Nm, Sagittal Velocity = 10 deg/sec (isokinetic))

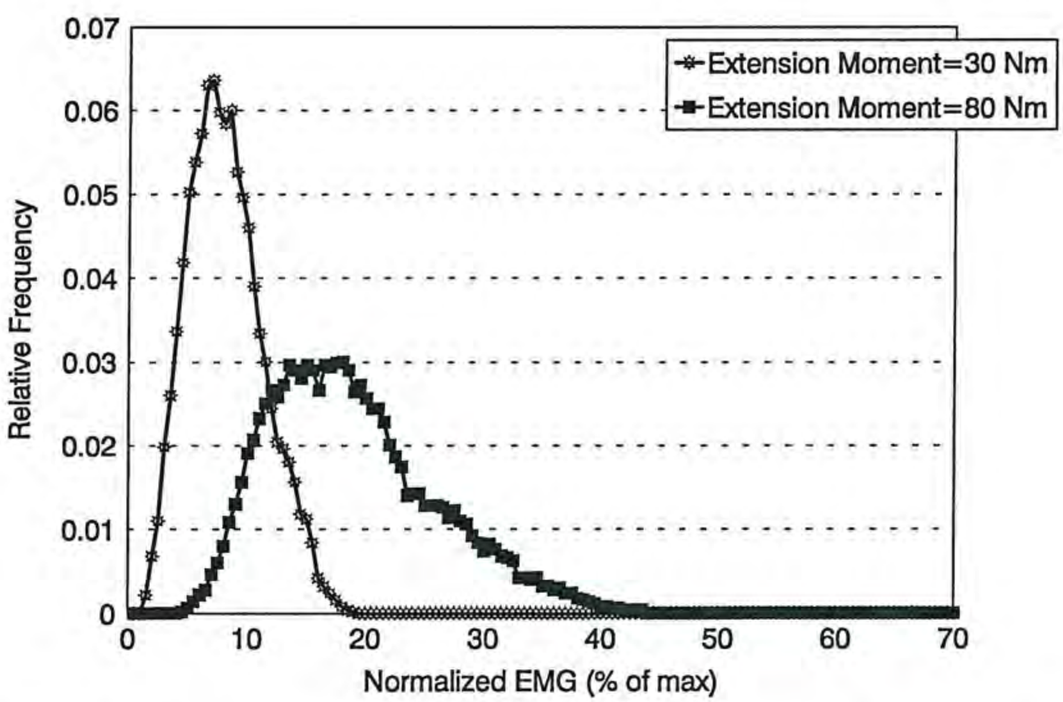

Figure 10. Distributions of the right latissimus dorsi as a function of extension moment. (Sagittal Angle = 5, Sagittal Velocity = 10 deg/sec (isokinetic), Sagittally Symmetric)

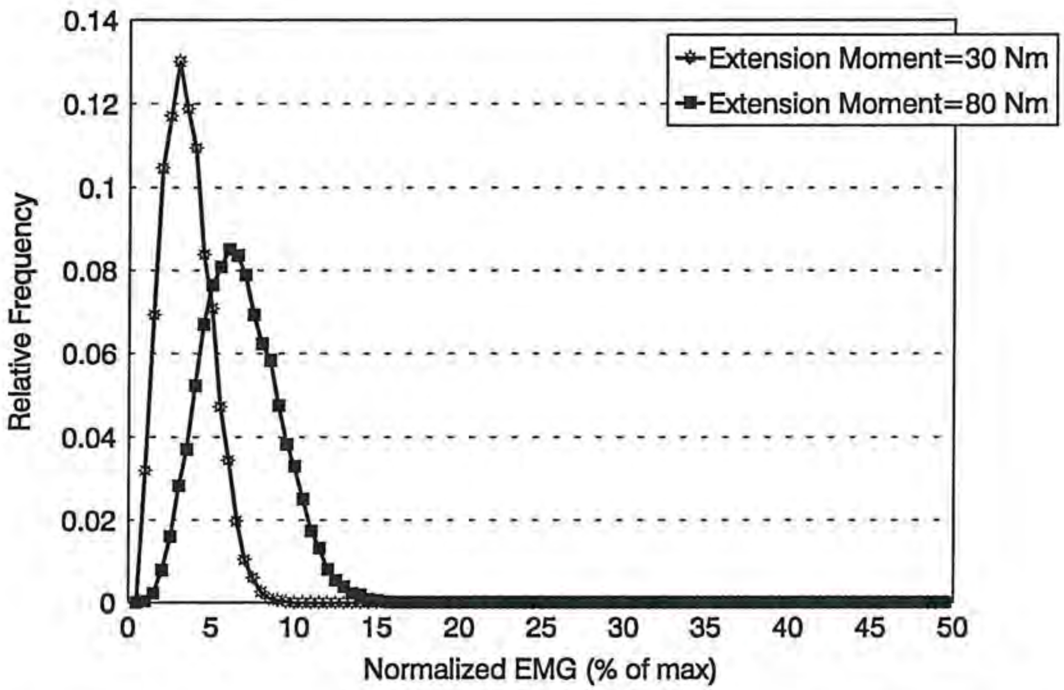

Figure 11. Distributions of the right external oblique as a function of extension moment. (Sagittal Angle = 5, Sagittal Velocity = 10 deg/sec (isokinetic), Sagittally Symmetric)

DISCUSSION

Many of the biomechanical models of the torso that have been developed rely on electromyographic inputs to drive the model (Granata and Marras, 1993; Marras and Sommerich, 1991a,b; McGill and Norman, 1986; McGill, 1992; Reilly and Marras, 1989). However, it is not practical to collect EMG activity in many industrial environments. The model developed in this research has the capability of generating muscle activities during bending and lifting activities and can therefore act as an engine for these EMG driven biomechanical models. Given a set of environmental conditions (load lifted, trunk posture and trunk dynamics) this model can produce EMG signals that would be generated during these exertions. These signals could then be input into any of the existing EMG driven model to render estimates of the spine reaction forces.

Variability in biomechanical systems is not a topic that has received much attention in the ergonomics literature. A couple of epidemiologic studies however, have indicated that there is a need to better understand the variability in biomechanical stresses placed on the worker to better understand risk of injury. Herrin et al (1986) performed detailed biomechanical analyses of 55 industrial jobs which entailed a total of 2934 MMH tasks. The results of their analysis showed that, as opposed to the average or aggregate requirements of a job, it was the most stressful tasks within that job which were found to be most predictive of overexertion injury. An industrial surveillance study performed by Marras et al (1993) used a trunk motion monitoring device to capture the continuous kinematic parameters that describe three-dimensional lumbar motion during MMH tasks. Their results showed that the peak kinematic variables (such as peak velocity in the coronal plane) add critical predictive power to models of occupational low back injury.

These two studies indicate that by focusing our attention on the average stresses encountered during a task, valuable information pertaining to risk may be lost. Therefore, by quantifying the variability of human performance during manual materials handling tasks, valuable insight into workplace risk can be gained. These studies focused on the gross variability on human performance, while the current study emphasized a controlled lifting environment and illustrated the variability that exists in the muscle coactivation strategies. Both of these types of variability should be considered in order truly understand spinal loading.

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Project #2

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BACKGROUND

Forces exerted on the spine during manual materials handling (MMH) tasks can be classified into one of two categories: external forces and internal forces. The external forces are generated outside of the body and are primarily of gravitational and inertial origin. These forces include gravity's pull on the mass of the various body segments and the mass of the load being lifted. In addition to these gravitational forces, there are also inertial forces that result from the dynamic component of the lifting activity. The internal forces, on the other hand, are those forces that are exerted by the musculoskeletal system. These internal forces include trunk muscle forces which provide the moments needed during lifting, as well as the compression, shear and torsional forces which the spine must produce, being the fulcrum of the biomechanical system. To quantify both instantaneous and cumulative spine loading (and their negative impact on the structural integrity of the spine), it is necessary to have a clear understanding of all aspects of both the internal and external forces that act on the spine.

One of the facets of this biomechanical system which has yet to be fully understood is it's stochastic nature. The multidimensional, indeterminate nature of most biomechanical systems indicates that there are many ways a person can perform a movement or exertion - both from an internal and external perspective. This suggests that a more appropriate method of analyzing the biomechanical system is to use a stochastic modeling approach which allows for an accounting of this flexibility. This stochastic modeling approach can be employed in two different ways: modeling the variability in human performance during MMH activities (external perspective) and modeling variability in the trunk muscle forces and the spine reaction forces (internal perspective).

With regard to the variability of the internal biomechanical stresses, a recent study by Mirka and Marras [11] revealed that there is a significant amount of variability in the internal stresses on the spine during lifting motions. In this laboratory study, subjects performed multiple repetitions of controlled trunk extension exertions within a reference frame which precisely controlled the kinematics of the lifting motion. As the subjects performed these simulated lifting exertions, the electromyographic (EMG) activities of ten trunk muscles were collected. The EMG data collected from identical lifting conditions were then combined, and distributions describing the activation levels of each of the ten trunk muscles were developed. The results of this study showed that there was significant variance in the muscle coactivation patterns employed during these relatively simple bending motions. It was further shown that the magnitude of this variability was affected by workplace parameters such as the moment exerted and the angular velocity of the lifting motion. These data were further examined using an EMG-driven biomechanical model [7] and the results of this analysis revealed that spine compression forces at three standard deviations above the mean were 7% greater than the mean spine compression force. This result was shown to be a function of the variable activation of the trunk musculature, particularly the erector spinae muscles. It was further revealed that this erector spinae variability affected, through complex coactivation patterns, the shearing forces experienced by the spine. Analysis of the shear forces revealed that the anterior/posterior shear forces at three standard deviations above the mean were 40% greater than the mean anterior/posterior shear force, while the lateral shear forces at this level were 90% greater than the mean lateral shear force. This study illustrated the potential variability of the internal forces and stressed the need to investigate the stochastic nature of all aspects of MMH tasks.

Based on some recent industrial surveillance studies, a measure of the variability of human performance during MMH activities may also contain valuable information towards the description of the etiology of low back injuries. Herrin et al [4] performed detailed biomechanical analyses of 55 industrial jobs which entailed a total of 2934 MMH tasks. The results of their analysis showed that, as opposed to the average or aggregate requirements of a job, it was the most stressful tasks within that job which were found to be most predictive of overexertion injury. An industrial surveillance study performed by Marras et al [6] used a trunk motion monitoring device to capture the continuous kinematic parameters that describe three-dimensional lumbar motion during MMH tasks. Their results showed that the peak kinematic variables (such as peak velocity in the coronal plane) add critical predictive power to models of occupational low back injury. These two studies indicate that by focusing our attention on the average stresses encountered during a task, valuable information pertaining to risk may be lost. Therefore, by quantifying the variability of human performance during manual materials handling tasks, valuable insight into workplace risk can be gained.

SPECIFIC AIMS

The specific aims of the present study are three-fold. First, quantify the variability of kinematic parameters describing human performance of simple sagittally symmetric MMH activities. Second, describe how this variable kinematic data translates into variable moment about the L5/S1 joint in the sagittal plane. Finally, investigate the effect of workplace variables on the magnitude of this variability.

METHODOLOGY

Subjects

Seven male college graduate and undergraduate students served as subjects in this experiment. None had a history of low back impairment/disorder. Subject population had a mean age of 25 years (sd = 2.98 years), a mean height of 179.0 cm (sd = 7.6 cm), and a mean mass of 82.9 kg (sd = 6.3 kg).

Experimental Design

Independent Variables. The independent variables in this study were the magnitude of the load and the quality of the coupling interface between the subject and the box being lifted. The levels of load magnitude were: 4.5, 9, 13.5, 18, 22.5, 27, and 31.5 kg. The coupling levels were good, fair and poor as described by the NIOSH revised lifting guidelines [16]. All lifts were performed in sagittally symmetric postures.

Dependent Variables. The dependent variables were the kinematic parameters describing the angular position, angular velocity and angular acceleration of the lumbar trunk in the sagittal, coronal and transverse planes. Subsequent calculations using a dynamic biomechanical model of the torso allowed for the calculation of a tenth dependent measure: moment in the sagittal plane about the L5/S1 joint.

Design. Each combination of independent variables was repeated 8 times per subject. The presentation of coupling conditions was randomized within weight levels while the load levels began at 4.5 kg and then increased in 4.5 kg increments. The levels of load were not randomized because data from a pilot study revealed that varying the weight randomly from trial to trial artificially increased the variability of the kinematic data because it took the subjects much longer to become comfortable with the weight lifted, resulting in an increase in the time it took them to reach steady state. The pilot study also revealed that concerns about the potential confounding effects of fatigue were minimized if the subjects were given adequate rest breaks between consecutive trials.

Apparatus

The motion of the lumbar region of the trunk was monitored using a device called the Lumbar Motion Monitor (LMM). This device was secured to the subject's back and measured the angular position of the lumbar spine in the sagittal, coronal and transverse planes. These position signals were then differentiated in software to get angular velocity and angular acceleration in the three cardinal planes. The three-dimensional trunk position data were collected at a rate of 60Hz. For a more complete description of the LMM and the processing of the data, see [5].

Subjects were asked to lift a 35cm x 35cm x 30cm wooden box that weighed 4.5 kg. Cast iron weights of 4.5 kg were added to create the designated load levels. In the good coupling condition, the subjects used cut-out handles on the box, which when used, placed the subjects metacarpophalangeal (MP) joint at 25 cm from the floor. (See Figure 1.) In the fair condition, the box was placed on a 25 cm stool to control for the distance the subjects had to stoop in order to reach the box. The subjects were instructed to lift the box from underneath, thus allowing the fingers to be flexed at 90 degrees. In the poor coupling condition the subjects were instructed to the lift the box with a compression type hold on the sides of the box. Care was taken to ensure that across each of the three coupling conditions the height of the MP joint above the ground as the subject grasped the box was a constant 25 cm.

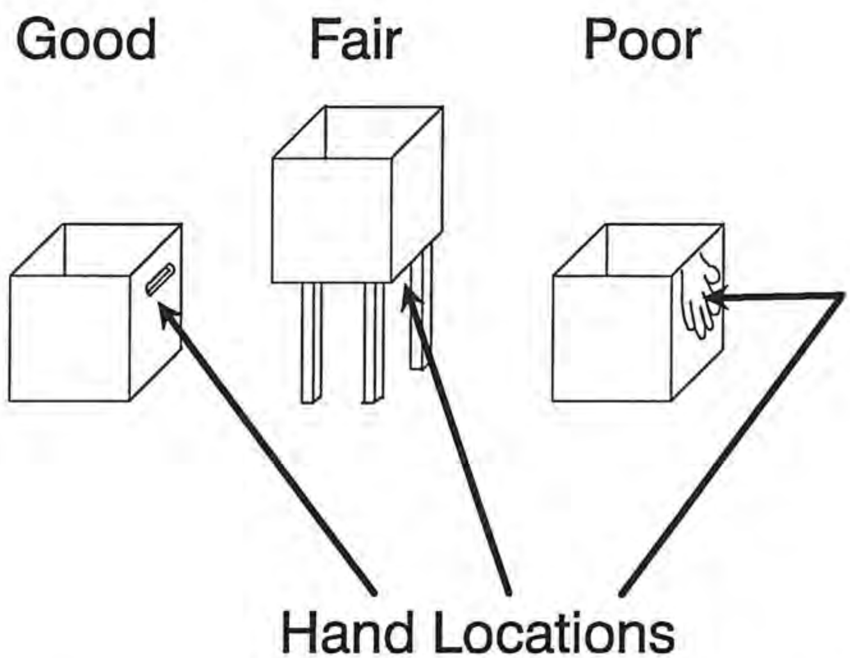

Figure 1. Illustration of the hand placement locations in each of the three coupling conditions. Good - use handles. Fair - lift box from underneath. Poor - compression lift on sides of box.

Procedure

Subjects began the experimental session with a brief warm-up and collection of anthropometric variables. They were then fitted with the LMM and were encouraged to move through a complete 3-D range of motion to assure themselves that the apparatus would not inhibit normal motion.. Baseline trunk position readings were collected as the subjects stood erect (sagittal angle = 0°) and in a sagittally symmetric 90° forward bend posture. These calibration data values were used to normalize the data during data processing.

Once these preliminary trials were complete, the subject moved to the lifting area. The weight of the first trial was 4.5 kg and the coupling was randomly set. The lift rate was set at 4 lifts per minute and the duration was three minutes for each experimental condition. The data from the first minute was discarded to allow for the subject to reach steady state, rendering a total of eight repetitions per experimental condition. The subjects were asked to lift the box "using the maximum comfortable speed that you would normally lift an object of this weight" while keeping their elbows and knees at a comfortable degree of flexion throughout the range of lifting motion. Most subjects chose to lift the box with straight arms and a slight angle of flexion in the knees. After completing this first set of eight lifts, the subjects rested for three minutes while the box was prepared for the next trial. This process continued until all of the trials within the capability of the subject were completed.

Data Processing

The results of the data collection revealed that the trials wherein the load was greater than 22.5 kg while the coupling quality was 'poor' were outside of the ability of most of the subjects. The compression force required was simply too great for all but one of the subjects. Therefore these two combinations were excluded from further analysis. All other combinations of coupling and load had at least 50 usable lifting trials (out a possible 56 (7 subjects x 8 repetitions per subject)).

The first step in data processing was to standardize the data by eliminating those data points which occurred before the subject's hands came in contact with the box. In each plane of the body, the range of motion, peak angular velocity and peak angular acceleration were then obtained for the remaining points which constituted the concentric portion of the lifting motion. Finally, the kinematic data from the LMM were input into a dynamic biomechanical model so that the peak moment about the lumbosacral joint could be calculated.

The biomechanical model calculated the time dependent, dynamic external moment about the lumbosacral joint. The body was partitioned into a 5-link system consisting of two upper extremity links, two lower extremity links, and a composite head/neck/trunk link beginning at the lumbosacral joint. The static anthropometry data of the each of the subjects were combined with data and regression equations from [1][3][13][14] to obtain estimates of 1) the trunk mass, 2) the distance between center of mass of trunk and L5/S1, 3) the mass of the arms, 4) the distance between the gleno-humeral joint center of rotation and L5/S1.

The time dependent sagittal position data was first normalized with respect to the 0° and 90° calibration values that were collected at the beginning of the experiment. This was done to control for variable lumbar lordosis between subjects. The normalized LMM data and the anthropometric values were then used as inputs to the dynamic biomechanical model which calculated the time dependent external moment in the sagittal plane about the L5/S1 intervertebral joint. The equation used to calculate the sagittal moment about L5/S1 is shown below (1). This equation was used to calculate the moment about L5/S1 in the sagittal plane at each instant in time at a frequency of sixty data points per second.

$$M(L5/S1) = MT*g*R1*\sin\theta + MAB*g*R2*\sin\theta + MAB*R2^2*\sin\theta*\alpha + I\alpha \qquad (1)$$

<u>Variable Definitions</u>

MT = mass of trunk (kg)
g = gravitational constant (9.81 m/sec/sec)
R1 = distance from L5/S1 to COG of trunk (m)
θ= sagittal angle of the trunk (upright = 0 deg)
MAB = mass of the arms plus mass of box and contents (kg)
R2 = distance from L5/S1 to gleno-humeral joint (m)
α = angular acceleration of trunk in the sagittal plane (rad/sec/sec)
I = mass moment of inertia of the trunk and head about L5/S1 (kg m^2)

Note that the third term in the above equation is describing a vertical linear acceleration of the load and arms which omits any description of horizontal inertial forces. This model is therefore only applicable in the early phase of the lifting motion when the horizontal motion of the shoulder joint is negligible. In order verify that this is an appropriate assumption for the current experiment, subjects stood on a force platform during lifting. The force platform data was used to verify that the peak vertical ground reaction forces (and thereby the peak moment values) occurred before significant horizontal ground reaction forces were developed.

RESULTS

Looking at the data qualitatively, Figures 2 and 3 illustrate the time dependent response of the angular position, angular velocity and angular acceleration in the sagittal plane. In these figures, time t=0 corresponds to that point during the lift when the subject's hands first touched the box. Note the differences in the variability between these two charts- particularly in the plot of acceleration. These figures further illustrate that, in addition to the variability in the magnitude of the peak value, there is also some variability in the location in time when this peak value occurs.

The results of a more quantitative analysis of trunk kinematics showed a significant amount of motion in all three planes. Table 1 shows the mean and standard deviation of the sagittal plane kinematic parameters as a function of the experimental conditions. It should be noted that these values do not represent the between-subject variability. The data was standardized so that the values in Table 1 reflect a pooled average of the subjects' variability within that particular lifting condition. In an effort to find the source of the variability, a repeated measures analysis of variance was performed to see if there were any consistent trends across the eight repetitions within each condition. This analysis, however, did not reveal any consistent trends across repetitions. Comparing the three levels of kinematic data, these results show that the higher derivatives of motion become relatively more variable. The average coefficient of variation (CV) for the range of motion in the sagittal plane was 2.5% while the CV for peak velocity was 8.1% and the CV for peak acceleration was 11.4%.

Kinematic results from the transverse and coronal planes also revealed a significant amount of dynamic activity in these "off-planes". These results are shown in Table 2. These kinematic parameters did not, however, show statistically significant trends as a function of load or quality of coupling.

From an ergonomic point of view the greatest impact of kinematic variability is in its effect on trunk kinetics. These effects were shown when the trunk motion data was input into the dynamic biomechanical model. The results of this analysis are shown in Table 3. Numerically, consideration of the range of the peak moments across trials shows that a moment at two standard deviations above the average is between 5% and 11.3% higher than the average of the peak sagittal moments. This illustrates that the peak moment occurring on what might be considered an average lifting exertion may not be representative of the type of loading which *could* occur under the same set of conditions simply due to changes in the lifting dynamics chosen by the subject.

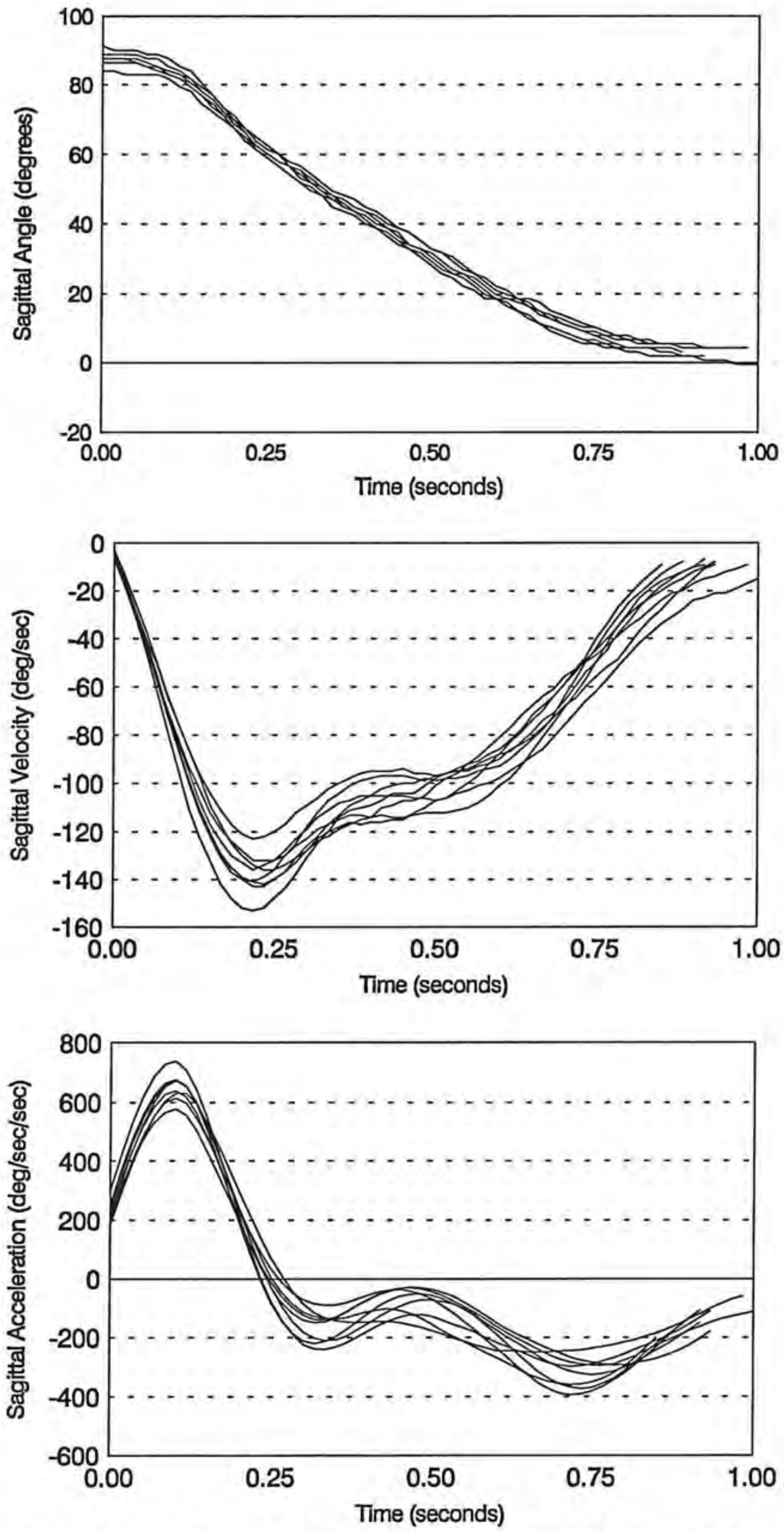

Figure 2. Time dependent traces of sagittal angle, sagittal velocity and sagittal acceleration. Load = 22.5 kg. Coupling = 'Good'

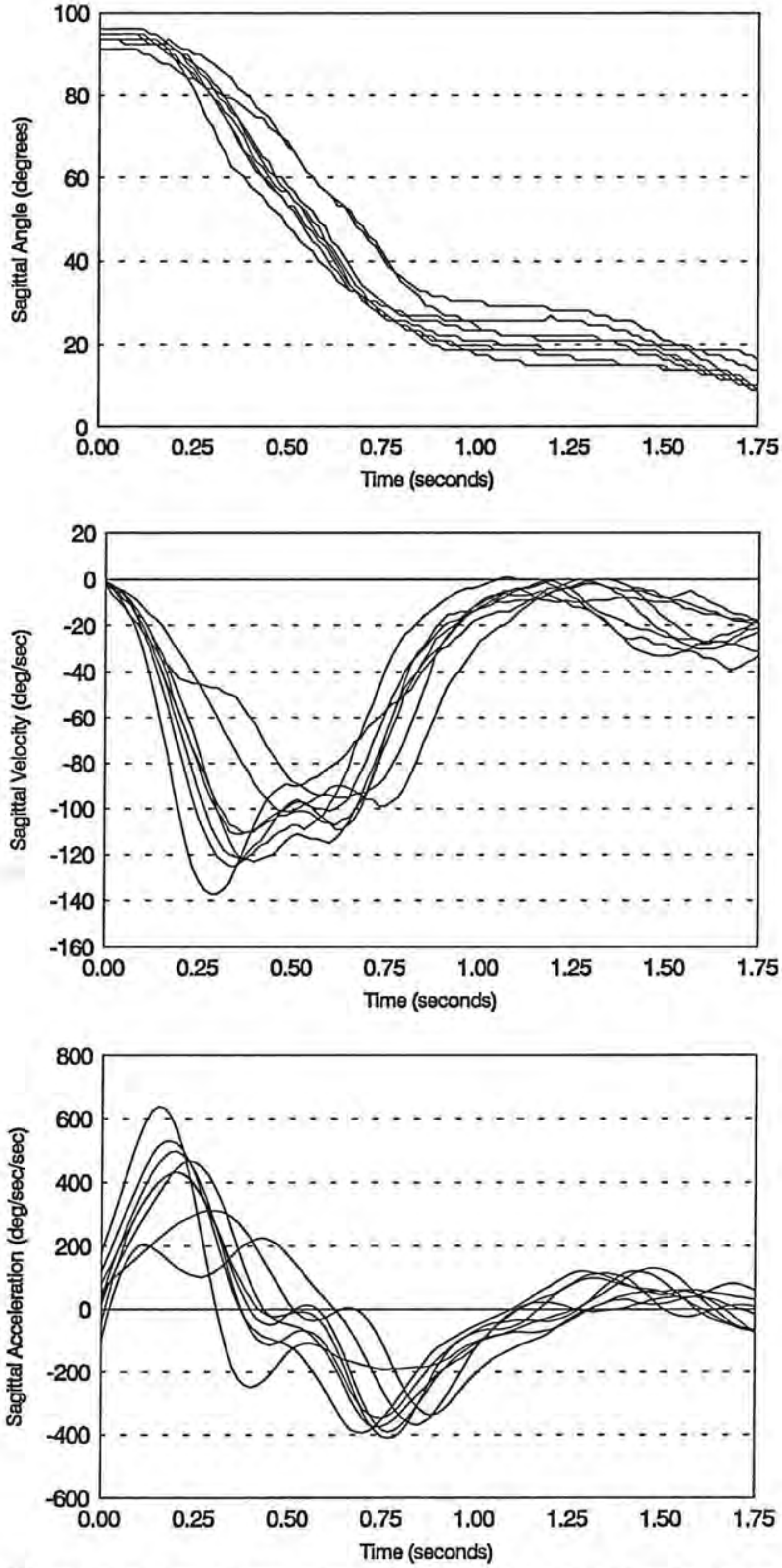

Figure 3. Time dependent traces of sagittal position, sagittal velocity and sagittal acceleration. Load = 22.5 kg. Coupling = 'Poor'

Table 1. Mean and standard deviation of peak kinematic parameters in the sagittal plane

LOAD(kg)		Quality of Hand to Container Coupling					
		'GOOD'		'FAIR'		'POOR'	
		Mean	STD	Mean	STD	Mean	STD
4.5	ANG (deg)	87.6	1.8	90.8	1.3	92.7	1.1
	VEL (deg/sec)	-165.2	11.4	-172.5	8.6	-166.6	9.5
	ACC (deg/sec^2)	745.3	57.4	746.3	57.1	710.2	53.5
9.0	ANG (deg)	87.4	2.6	88.8	2.0	92.3	2.2
	VEL (deg/sec)	-166.6	12.9	-162.5	9.6	-166.6	10.0
	ACC (deg/sec^2)	756.9	62.9	718.9	51.0	703.5	51.6
13.5	ANG (deg)	87.2	3.1	87.7	2.5	92.0	0.8
	VEL (deg/sec)	-158.9	11.9	-155.8	10.1	-154.3	10.3
	ACC (deg/sec^2)	699.5	63.5	685.4	50.8	643.3	59.8
18.0	ANG (deg)	87.2	2.9	87.9	3.3	91.6	1.2
	VEL (deg/sec)	-153.0	10.0	-149.9	11.2	-137.1	10.6
	ACC (deg/sec^2)	667.4	68.4	666.7	60.2	547.6	60.8
22.5	ANG (deg)	85.2	2.2	86.7	2.7	92.8	1.8
	VEL (deg/sec)	-137.5	13.0	-137.3	8.2	-119.3	16.1
	ACC (deg/sec^2)	592.2	73.1	604.2	49.1	415.6	95.8
27.0	ANG (deg)	84.8	3.1	84.5	1.9		
	VEL (deg/sec)	-131.8	13.8	-122.9	8.8		
	ACC (deg/sec^2)	563.7	83.0	521.9	53.7		
31.5	ANG (deg)	85.2	2.6	86.1	1.6		
	VEL (deg/sec)	-124.3	12.4	-121.1	8.0		
	ACC (deg/sec^2)	504.7	56.4	485.5	43.7		

Table 2. Mean and standard deviation of the peak kinematic parameters in the coronal and transverse planes

	MEAN	STD
Coronal Range of Motion	4.6 (deg)	1.7 (deg)
Transverse Range of Motion	3.1 (deg)	1.7 (deg)
Max Coronal Velocity	13.6 (deg/sec)	4.6 (deg/sec)
Max Transverse Velocity	7.8 (deg/sec)	3.1 (deg/sec)
Max Coronal Acceleration	63.2 (deg/sec^2)	22.6 (deg/sec^2)
Max Transverse Acceleration	40.4 (deg/sec^2)	13.7 (deg/sec^2)

Table 3. Mean and standard deviation of peak moment in the sagittal plane (in Nm)

	Quality of Hand to Container Coupling					
LOAD(kg)	'GOOD'		'FAIR'		'POOR'	
	Mean	STD	Mean	STD	Mean	STD
4.5	268.9	7.4	270.8	7.6	265.8	6.3
9.0	299.5	9.1	294.3	7.4	292.2	7.3
13.5	317.8	10.5	316.6	8.6	310.2	9.5
18.0	340.4	12.7	340.6	11.1	321.5	10.9
22.5	353.5	13.9	356.8	10.3	319.4	18.1
27.0	375.2	17.5	366.2	11.3		
31.5	390.1	12.3	386.3	10.0		

In an attempt to understand the effect of workplace factors on the magnitude of this variability, differences in the variance of the peak moments were tested using the Bartlett test for equality of variances. The first step in this analysis process was to test the assumption of normality of the data in each of the experimental conditions. The Shapiro-Wilk statistic (W) was computed for each of the combinations of independent variables to test the null hypothesis that the data from each of these cells was were from a normal distribution. The lowest W found was .91 and the average across cells was .96, indicating that these data fit the normal distribution. Due to the unbalanced nature of the data set, the data set analyzed using the Bartlett test differed depending on the effects to be tested. To test the effects of load magnitude on the amount of variability, only the data from the good and fair coupling conditions was analyzed. These results showed a significant effect ($B = 47.75$; $\chi^2_{(.05,6)} = 12.592$) due to the load weight, with greater weight levels showing a higher level of variability. This significant weight effect is graphically illustrated in Figure 4. (The distributions shown in this figure are parameterized Johnson distributions which have been fit to the empirical data of this experiment. See [2][11] for further details on fitting empirical data to the Johnson family of distributions.) To test for a coupling main effect, only the load levels which were less than 27 kg were considered. The Bartlett test for equality of variance on this data revealed that coupling as a main effect had no significant effect on the magnitude of the variance.

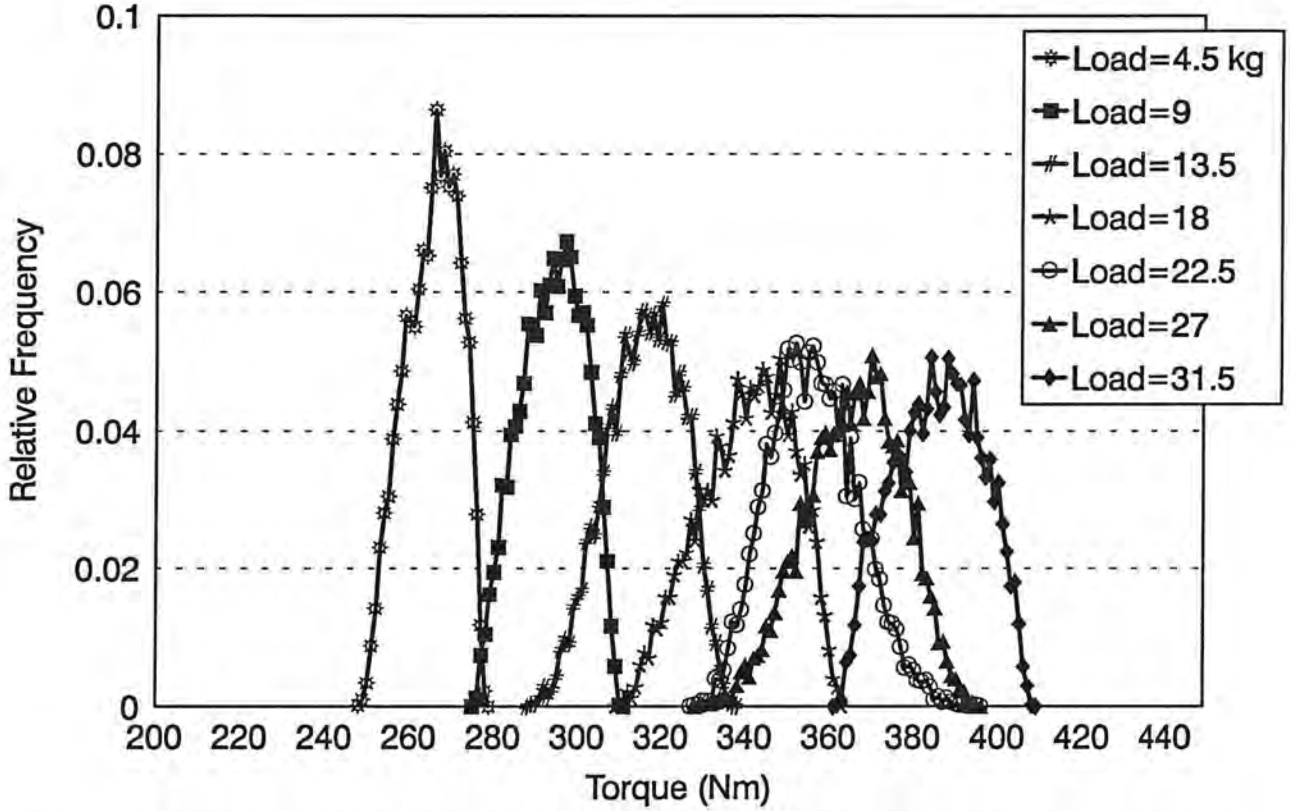

Figure 4. Distributions of peak sagittal moment as a function of load magnitude. Note that in addition to the increase in the mean at higher load levels, there is also an increase in the variability.

An analysis of the response of the means showed that load level had a significant ($p<.0001$) effect on the magnitude of the peak moment, while coupling effects were shown to only be significant under specific load levels. To make the data collected in this study useful to individuals wishing to use this data in a stochastic model of human performance, regression analysis was performed on the peak sagittal moment data. Variables considered for this equation included: subject weight, subject height, load and $\{\text{load}\}^2$. The following equations were the best fit the empirical data.

$$\textbf{Mean (Nm)} = 248.47 + 5.023 * \text{Load (kg)} - .0220 * \{\text{Load (kg)}\}^2 \quad (R^2 = .874) \tag{2}$$

$$\textbf{Std Dev (Nm)} = 1.16 * \text{Load (kg)} + -.0238 * \{\text{Load (kg)}\}^2 \quad (R^2 = .961) \tag{3}$$

DISCUSSION

Variability in the biomechanical aspects of human performance is a research topic which has not received much emphasis in the industrial ergonomics literature. However, as models and standards begin to reflect the addition of dynamic effects, they must also begin to reflect the human performance aspect of lifting. The total biomechanical stress associated with a dynamic lifting task is not limited to simple workplace dimensional information but now can be affected by the lifter himself. When static models of the trunk during lifting were considered adequate to represent lumbar stress, the human performance component could be easily overlooked. Muscle co-contraction/antagonism were considered negligible thus making equilibrium between internal and external forces relatively easy to calculate. However, as models begin to consider dynamic effects, the assumption of insignificant muscle antagonism/co-contraction is brought into question (Marras and Mirka [8][9][10]). Further, these models must also address human performance issues because the human has the ability to alter the dynamics of the system by deciding an appropriate velocity/acceleration to be used to perform the task. Given that the lifter has the ability to change the lifting method, variability in this method or strategy will lead to variability in the biomechanical stresses experienced by the spine and ultimately the risk of low back injury. Stochastic models are uniquely qualified to describe this risk because 1) they are capable of describing the magnitude of variable peak loads, 2) they are capable of quantifying the likelihood that these peak loads occur and 3) they may render a better understanding of the long term cumulative effect of lifting.

There has been some work which has shown how variability is affected by extreme lifting conditions. Previous work done by Parnianpour et al [12] showed significant motion in the coronal and transverse planes as a function of fatigue. In this previous study, subjects performed repeated trunk flexion and extension exertions against a resistance equal to 70% of the subject's maximum isometric trunk extension strength. Their results showed a significant increase in the motion in the transverse and coronal planes as the subject became more fatigued. The results of the present study have illustrated that there is significant motion in these off-planes even under unfatigued conditions- a result which indicates that current research may be over-simplifying spinal loading by considering primarily the compression forces in the spine to the exclusion of the more complex, coupled loads which have been noted for their role in low back injury [15].

When the human performance variability demonstrated in this study is combined with the variability in internal forces shown by Mirka and Marras [11], it becomes evident that in order to accurately quantify spine loading and risk of low-back injury, a clear representation of the stochastic nature of the system is required. This study has shown that by considering only the mean or average performance profile, the peak moment is underestimated by between 5% and 11%, for one lift out of every fifty. This was the result in an experimental environment where many factors found on a plant floor which could potentially increase variability were eliminated. Thus, the results of this experiment should be considered the best case scenario because in more realistic workplace conditions, there will be environmental distracters which will tend to increase this variability in performance.

Future work in this area will be to find consistent patterns in the response of the mean and variance of human performance parameters as a function of workplace variables, and then to develop a complete stochastic model of human performance using this data. The regression equations developed in this research can form the basis of a relatively simple stochastic model of human performance during lifting. The data collected in this study was found to be a reasonable fit to normal distributions. Using the regression equations (2) and (3) one can simulate the cumulative loading by sampling repetitively from these distributions. It was interesting to note that although they were considered, subject anthropometric variables did not significantly add to the predictive power of the regression analysis.

Quite often in traditional quality control applications, industrial engineers are just as concerned with variability about a mean as they are with the mean level of quality itself. It is believed that this analogy can be carried into the area of ergonomics. Given a workspace and two alternative designs, it is conceivable that the best choice to reduce risk of low back injury may be that configuration which on average generates the higher internal loads; but because the variance about the mean in this configuration is lower, the risk of extreme loads, and ultimately risk of low back injury, is reduced. As this area of ergonomics begins to develop quantitatively, ergonomists may begin to concern themselves with variance reduction/control procedures when designing workspaces to reduce musculoskeletal injuries.

CONCLUSIONS

The goal of this study was to obtain a quantitative description of the magnitude of the variability of the kinematic and kinetic parameters that describe the human performance of lifting tasks. Tables 1-3 display the results of this research in numerical form. It has been shown that workplace variables do have an effect on the magnitude of this variability and it is proposed that these data can be used as building blocks for the development of more complex stochastic models of human performance which may be will better able to predict risk of injury for occupational manual material handling tasks.

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Project #3

Publications

Mirka and Kelaher, "The Effects of Lifting Frequency on the Dynamics of Lifting", Proceedings of the Human Factors and Ergonomics Society 39th Annual Meeting, pp 550-554.

BACKGROUND

The frequency multiplier found in the NIOSH equation (Waters et al, 1993) is based on data from psychophysical studies (Snook and Ciriello, 1991) and physiological studies (Garg, 1976, Garg et al, 1978). The psychophysical approach was used to develop the multiplier for frequencies up to four lifts/minute while the physiological results were used to find the multipliers for frequencies greater than four lifts/minute. These studies were useful in predicting preferred workload of an individual (psychophysical) or the cardiovascular cost of lifting (physiological), but they did not discuss the biomechanical implications of variable lifting frequency. Given that frequency of lift does not fit well into a static description of a work environment, trunk kinematics during manual lifting tasks need to be considered in order to get a more complete picture of the risk associated with frequency of lift in free dynamic MMH tasks. The goal of the current study was to better understand the effects of lifting frequency on these trunk kinematic parameters.

SPECIFIC AIMS

The goal of this study was to quantify the effects of different lifting frequencies (3, 6 and 9 lifts/minute) at different lifting heights (30 and 60 cm) on the kinematics of the lumbar region. Each of these lifting tasks was performed for twenty minutes. The time dependent traces of the both the mean and standard deviation of sagittal acceleration showed subject dependent trends over time. Averaged across time, the results of this study reveal that there is a non-linear increase in the sagittal acceleration with greater frequency of lifting.

METHODOLOGY

Subjects

Five male and five female college students volunteered for this study. The subjects had no history of a low back disorder. Written informed consent was given by all of the subjects at the beginning of the experiment.

Equipment

Motion analysis and heart rate monitoring equipment were utilized for this experiment. A telemetry-based Lumbar Motion Monitor was placed on the back of the individual to measure the angular position, velocity, and acceleration of the lumbar spine in three-dimensional space. The heart rate of the subjects was measured using an Accurex II Heart Rate Monitor.

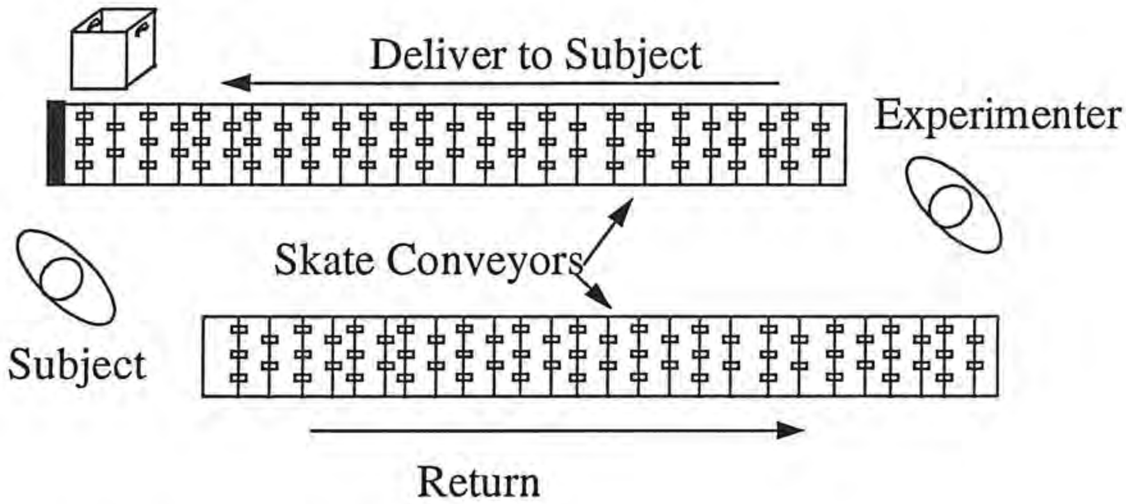

Figure 1. Overhead view of experimental environment

Experimental Setup and Task

Wooden boxes with handles ("good" coupling as defined by the 1991 Lifting Equation) were delivered to the subjects via gravity-fed passive skate-wheel conveyers at approximately 5° slope. The mass of the boxes were 11.4 kg for men and 7.4 kg for the women. The conveyors were placed parallel to each other 2 feet apart and were offset by 2 feet. This resulted in a 90° rotation requirement for transferring the boxes from the first to the second conveyer (Figure 1). Subjects worked for twenty minutes and then were given a five minute rest period.

The boxes were placed on the conveyer at the prescribed frequencies to simulate equally spaced packages from an automatic conveyer system. Each box was sent to the subject with the handles in line with conveyor to standardize the initial lifting posture. The subjects were then asked to lift the box after it reached the box stop and place it on the top of the return conveyer with the handles perpendicular to the midline of the conveyor. The subjects were given no instructions as to the lifting style to be used. Research has shown that this free-style lift allows for the greatest psychophysical lifting capacity (Garg and Saxena, 1979) while providing data which would be most representative of industrial data.

Experimental Design

The independent variables for this study were frequency of lift and the height of the handles as the box was lifted. The starting heights were 30 and 60 cm from the floor, while the height of the handles while placing the box on the return conveyer was 70 cm. The three lifting frequencies chosen for this study were 3, 6, and 9 lifts/minute. These frequencies were chosen for comparison with the data from previous research (Garg and Saxena, 1979) and to cover the range of frequencies typically seen in industry (Marras et al, 1993). The duration of each experimental condition was twenty minutes. Data was collected on each of the lifts for the 3 and 6 lifts/minute condition while computer limitations only allowed for data collection on every other lift for the 9 lifts per minute condition

The dependent variables were the kinematic parameters describing the motion of the lumbar spine (range of motion, velocity, and acceleration) in the three planes (sagittal, coronal, and transverse). This kinematic data was collected at a rate of 60 samples per second. Heart rate was collected as the instantaneous value at one minute intervals. Sagittal acceleration is the only dependent variable which will be discussed in the current paper.

Data Analysis

The peak value of sagittal acceleration was extracted from each of the data files. This peak value was computed as the average of the peak value and one data point on each side of this peak value. This was done to reduce data processing induced variability. This resulted in one peak value per trial. The time dependent standard deviation was then calculated from seven consecutive trials.

RESULTS

The results of this study are shown graphically in Figures 2-5. Figure 2 shows the response of sagittal acceleration to the different levels of starting height and lift frequency. A statistical analysis revealed a significant effect for both starting height and lift frequency ($p<.001$). Figure 3 shows an example of a time dependent response of the peak sagittal acceleration. The data in this part of the analysis was highly variable from subject to subject. In fact when these were averaged across subjects no significant trends resulted. It was only when the data was plotted by subject did the analysis reveal these interesting results. Finally, Figures 4-5 shown how the variability in the peak sagittal acceleration changed as a function of time. Again these trends were highly subject dependent.

DISCUSSION

The results of the present study suggest that in addition to the increase in some of the physiologic parameters which have been shown previously, there are some critical biomechanical parameters which are affected by increased frequency of lifting. It should be noted that the frequency levels chosen in this experiment were not so high as to require continuous work from the subject. If the frequency levels chosen had been at these excessive levels, it would not have been surprising for the frequency levels to be highly correlated with some of these kinematic parameters because the subjects' trunk dynamics would have to increase to keep up with the incoming work. The results of this study have shown, however, that under the conditions which do not require continuous work there still was an increase in most kinematic parameters with increased lifting frequency. Even under the highest work rate (9 lifts per minute) there still was between 3 and 4 seconds of rest wherein the subject was standing erect and waiting for the next box to arrive. This result indicates that instead of a required increase in trunk kinematics, as would be the case under conditions of high lift rates (>15/minute), there appears to some other mechanism at work.

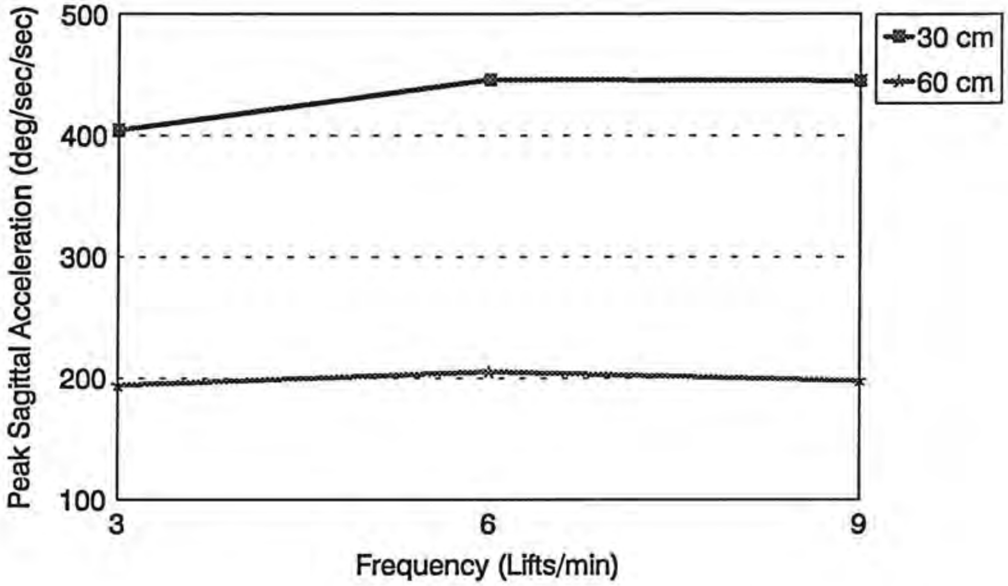

Figure 2. Response of sagittal acceleration to different levels of lifting frequency and starting heights.

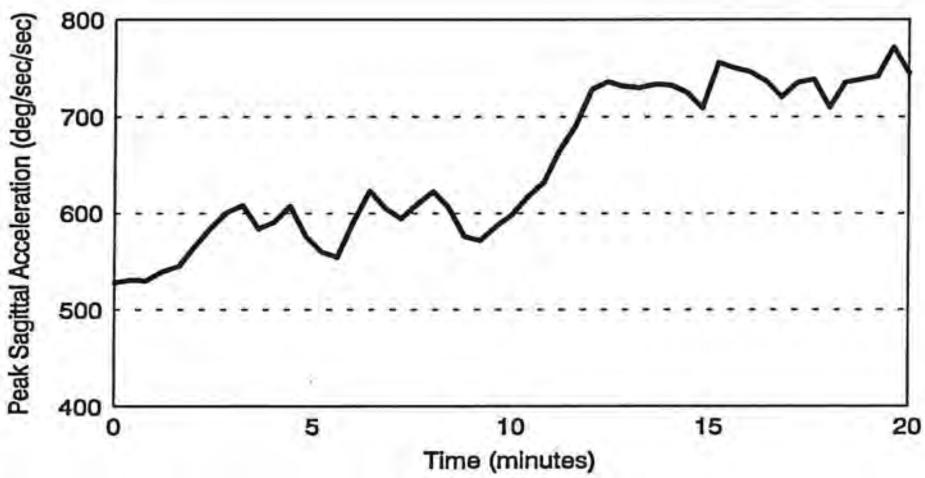

Figure 3. Time dependent response of the peak sagittal acceleration. Condition: Frequency = 9 lifts/minute, Starting Height = 30 cm.

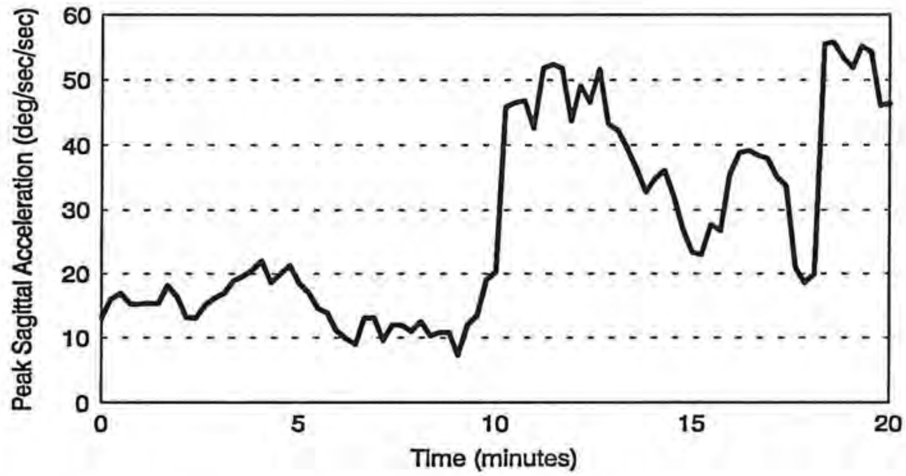

Figure 4. Time dependent response of the standard deviation of the peak sagittal acceleration. Condition: Frequency = 9 lifts/minute, Starting Height = 30 cm.

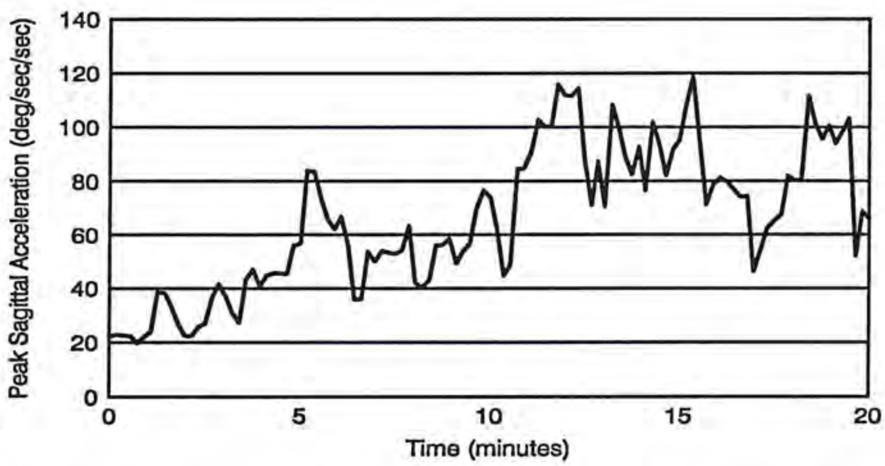

Figure 5. Time dependent response of the standard deviation of the peak sagittal acceleration. Condition: Frequency = 9 lifts/minute, Starting Height = 30 cm.

There has quite a bit of work done in the area of the effects of frequency during lifting. Much of the work has been in the area of psychophysics and cardiovascular physiology. Of particular relevance to the current study is a paper by Garg and Saxena (1979) which described a psychophysical study wherein subjects performed lifts at constant frequencies (3, 6, 9, and 12 lifts/minute) and were asked to find their maximum acceptable weight of lift. They found an interesting energy minimization at a frequency of 9 lifts/minute. The metabolic cost per unit work curve (i.e., Kcal/Kg*m or HR/Kg*m ratio) was parabolic with the minimum point occurring at the frequency of 9 lifts/minute (Garg and Saxena, 1979; Garg and Banaag, 1988). These studies were very useful in predicting the weights and workloads that subjects would choose as a function of the lifting frequency, but they did not discuss any change in the lifting biomechanics, a limitation which has been overcome somewhat with the current study. It is interesting to note that there does seem to be a sort of leveling off of the sagittal acceleration, a results consistent with that of the studies mentioned above.

The time dependent data shown in Figures 3-5 showed some very interesting subject dependent trends. Of particular interest is data presented in Figures 3 and 4. At about the ten minute mark there is an abrupt change in the variability of the subjects performance and an increase in the peak sagittal acceleration. Reviewing written notes taken during subject data collection it was noted that it was around this point that the subject changed from a squat lift to a stoop lift. This would explain the trends shown in these figures. Figure 5 shows a gradual increase in the variability of the data. This may be attributed to the gradual onset of fatigue. Not shown in this paper are data which show fairly high levels of variability early in the twenty minute period which tended to level off as the subject progressed through the experiment. This might indicate a short adjustment period as the subject became acclimated to the lifting task. Taken as a whole, the time dependent data suggests that it may be important for ergonomists to consider the changes over time of the human performance aspect of manual material handling. These parameters could become markers describing variables such as fatigue or warm-up periods which could be useful in setting up work schedules.

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Project #4

Publications

Research Proposal: Back Injury Interventions for Small Contractors, Grant#: 1 RO1 CCR 413061, US Department of Health and Human Services.

BACKGROUND

Current models used to quantify low back injury risk in traditional manufacturing industries can be placed into one of two categories: those that are best able to quantify acute trauma risk or those that are able to quantify cumulative trauma risk. The high degree of variability in biomechanical demands of many construction industry jobs illustrate the potential for both a cumulative and acute variety of back injury and therefore no single existing model is going to adequately describe overall risk. Preliminary data suggest that what is needed is a stochastic model that describes these variable biomechanical demands in such a way that a combination of existing models can adequately describe risk of injury.

The three models that will be considered in the proposed project are: 1) The 1993 NIOSH Lifting Equation, 2) University of Michigan's 3-D Static Strength Prediction Model, 3) The Lumbar Motion Monitor Model (Marras, 1993). Each of these models will be briefly outlined below.

The Revised NIOSH Lifting Equation (Waters et al, 1993) utilizes static workplace configuration information to develop estimates of the weights that can be safely lifted by a majority of the working population. The workplace variables considered in these models are: starting height of lift, ending height of lift, distance between the load and the spine, frequency of lifting, vertical travel distance of the load asymmetric posture of torso and coupling quality between the lifter and the object being lifted. These measures are then combined in a multiplicative model to arrive at an appropriate weight to be lifted given a specific workplace configuration (called the Recommended Weight Limit, or RWL). This RWL can then be compared to the actual weight being lifted to generate a value called the Lifting Index, which is the ratio of the actual weight to the RWL.

One of the limitations of this approach however is that it is based on static modeling and was developed to model jobs that are relatively repetitive in nature. A more recent model of low back injury was developed by Marras et al (1993). These researchers recognized that a static model of the workplace had only limited applicability in most workplaces and that the dynamics of lifting should be considered. Towards that end they utilized a device called the Lumbar Motion Monitor (LMM) that captured the instantaneous position, velocity and acceleration of the lumbar spine in three dimensions. Using this device in industrial environments they were able to develop a model of low back injury risk. To accomplish this they used multiple logistic regression to form a relationship between the historical injury data and a set of static and dynamic workplace parameters. The results of their analysis showed that a set of five parameters was adequate to distinguish between high and low risk jobs: lift rate, maximum sagittal angle, average twisting velocity, maximum lateral velocity and maximum moment. Using these five variables they developed a multiple logistic regression model of low back injury that had an odds ratio of 10.7, indicating that the model is almost 11 times more likely than chance to place a high risk job in the high risk category. The output from this model is a single value that describes the probability of high risk group membership. While this model was able to overcome the static modeling limitations of the NIOSH Lifting Guides, a limitation to the generalizabilty of this model is that it was developed from data collected while workers performed "... repetitive jobs without job rotation...". The model's validity, therefore, is limited to those types of jobs.

The 3-D Static Strength Prediction model (3DSSPP^{TM}), on the other hand is not limited to or based upon highly repetitive jobs. This model was based on a very large database of human strength capability that was collected by researchers at The University of Michigan. The input to this model is the posture of the series of joints running from the feet to the hands and the direction and magnitude of the force exerted by the hands. The output of the model is the percentage of the population that have adequate strength in each of the joints along this chain to be able to exert the designated force vector. A limitation of this approach is that we are documenting a one-time strength assessment and therefore have limited ability to quantify the cumulative risk, a factor that is better represented by the NIOSH and LMM models.

In summary, each of the above outlined models have strengths and limitations from the perspective of being able to quantify acute and cumulative trauma risk. When confronted with the task of assessing risk in an industry such as construction whose biomechanical demands can be so variable throughout the workday, it makes sense to combine features of each of these models and evaluate the impact that our interventions have on the distributions of biomechanical stress to which the workers are exposed.

METHODOLOGY

Subjects

Twenty individuals from each of three trades (masons, carpenters, roofers) will participate in this smaller, more descriptive study. The age, gender and race breakdown of these subject samples will reflect the proportion of each group as defined for each trade in the large sample study outlined above. Each of these subjects will have at least 6 months experience in their current jobs (not necessarily with their current employer) in an attempt to control for movement along their learning curve. Where possible the same subjects will be evaluated in all phases of this small sample study - pre-intervention evaluation, intervention implementation and post-intervention evaluation.

Instrument/Instrumentation

The instrumentation that will be used to assess the biomechanical stresses associated with the sampled jobs will include a portable, telemetry-based Lumbar Motion Monitor for collecting the kinematics of the torso (for use in the LMM model), a video camera for gathering the whole body postures (for input into the 3D Static Strength Prediction Program) and tape measures, scales and goniometers to get the static workplace dimensions (for gathering data to be used in the NIOSH Lifting Guide model). The LMM data will be collected on a portable 486 laptop computer with an analog to digital converter.

Data Collection Procedures

Each of the 60 subjects in this study will be followed for one complete workday. We will ask the individuals to arrive at the job site 45 minutes ahead of the start of the workday and begin by describing in general terms the study and their part in the study. They will then be asked to sign the informed consent form.

During the workday the research team will document the different manual materials handling tasks that the individual performed gathering data required to perform a complete evaluation using the three chosen assessment tools (Revised NIOSH Lifting Equation, LMM model and the 3DSSPP^{TM}). This data will include workplace dimensional information (starting heights, ending heights, weights of objects lifted, frequency of lifts, asymmetric postures, duration of static postures), lumbar kinematics data (position, velocity and acceleration of the lumbar spine in the sagittal, coronal and transverse planes) and video data. No restrictions will be placed on the individual as far as their workplace tasks. During breaks and lunch we will remove the LMM.

Data Analysis Procedures

The analysis of the biomechanical exposure data will consist of calculating three parameters from each manual materials handling task that the individual performs. This will include a Lifting Index from the NIOSH Lifting Guide, a Probability of High Risk Group Membership from the LMM model and a Percent Capable figure from the 3D Static Strength model. From this data exposure profiles in the form of histograms will be developed for each individual for each work day.

The distributions from both before and after interventions will be compared for each of the trades through the generation of parameters describing these distributions. These parameters will attempt to describe both the acute and cumulative trauma risk of the tasks. These parameters will include:

1) the percentage of lifts with a lifting index (NIOSH Lifting Guide) of greater than 5
2) the percentage of lifts with a lifting index (NIOSH Lifting Guide) of greater than 3
3) the percentage of lifts with a lifting index (NIOSH Lifting Guide) of greater than 2
4) the percentage of lifts with a lifting index (NIOSH Lifting Guide) of greater than 1
5) the percentage of lifts that have a probability of high risk group membership (LMM model) of greater than 50%
6) the percentage of lifts that have a probability of high risk group membership (LMM model) of greater than 75%
7) the percentage of lifts that have a probability of high risk group membership (LMM model) of greater than 90%
8) the percentage of lifts that have a probability of high risk group membership (LMM model) of greater than 98%
9) the percentage of lifts that where less than 50% of the people have the adequate strength capacity (3DSSPP^{TM})
10) the percentage of lifts that where less than 25% of the people have the adequate strength capacity (3DSSPP^{TM})

11) the percentage of lifts that where less than 10% of the people have the adequate strength capacity (3DSSPP™)
12) the percentage of lifts that where less than 2% of the people have the adequate strength capacity (3DSSPP™)

For each of the above 12 variables an analysis of variance procedure will be conducted in an attempt to find the effects of the ergonomic interventions.

PILOT RESULTS

This exposure assessment model was applied to several jobs in the construction industry: mason, mason's helper, drywall installer, mortar mixer and framers (carpentry work). This technique involves documenting the manual materials handling tasks that the subjects perform and then evaluating those tasks with known assessment tools. Figures 1-4 show the results of this analysis using the NIOSH Revised Lifting Equation as the assessment tool. Plotted on these figures are the number of lifts at the different levels of the Lifting Index. Note the differences between the different jobs with regard to the degree of variability present.

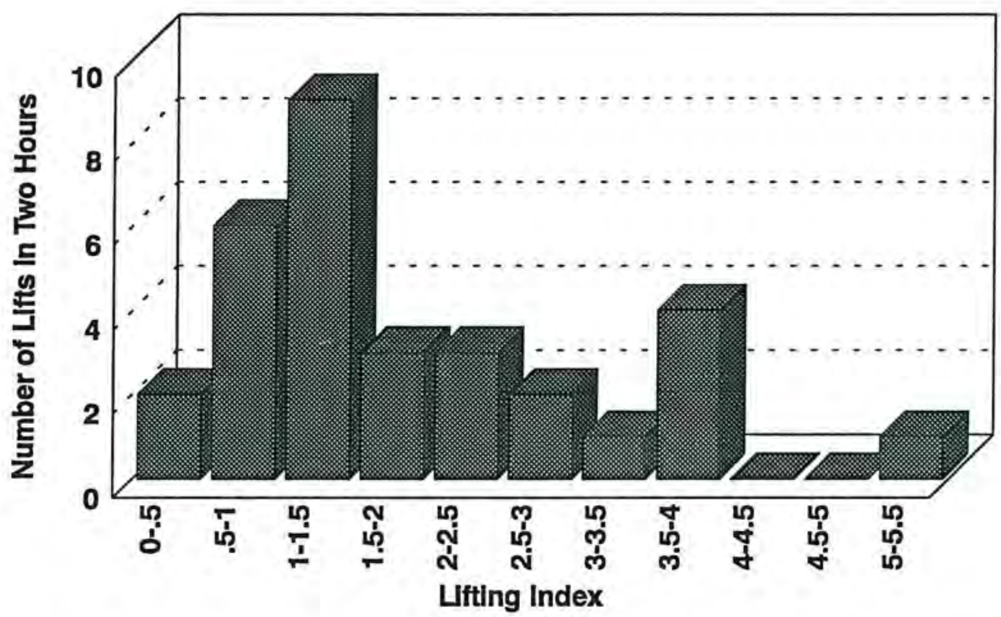

Figure 1. NIOSH Lifting Guide Assessment of a Sheet Rock Handler

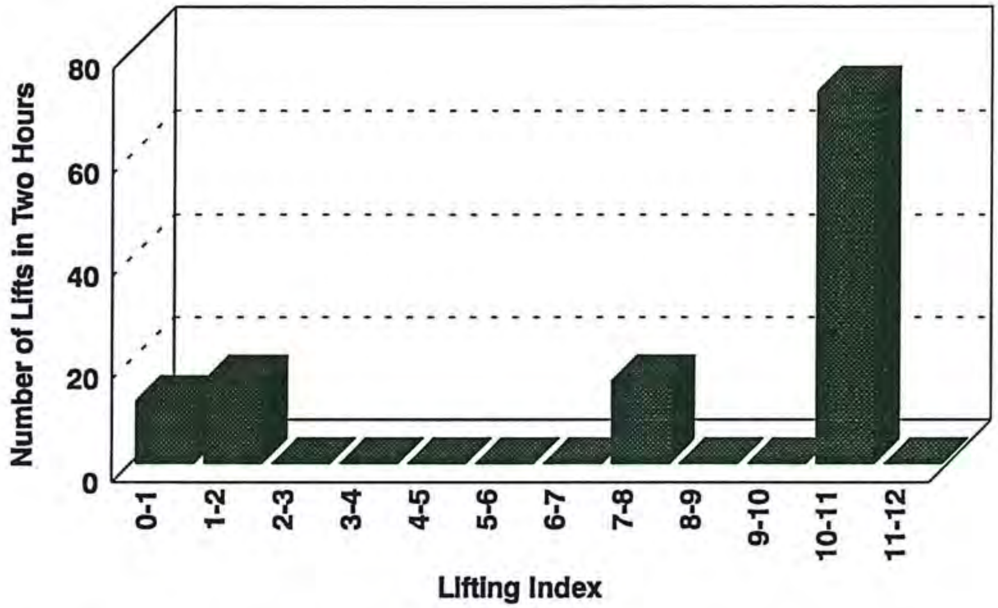

Figure 2. NIOSH Lifting Guide Assessment of a Mortar Mixer

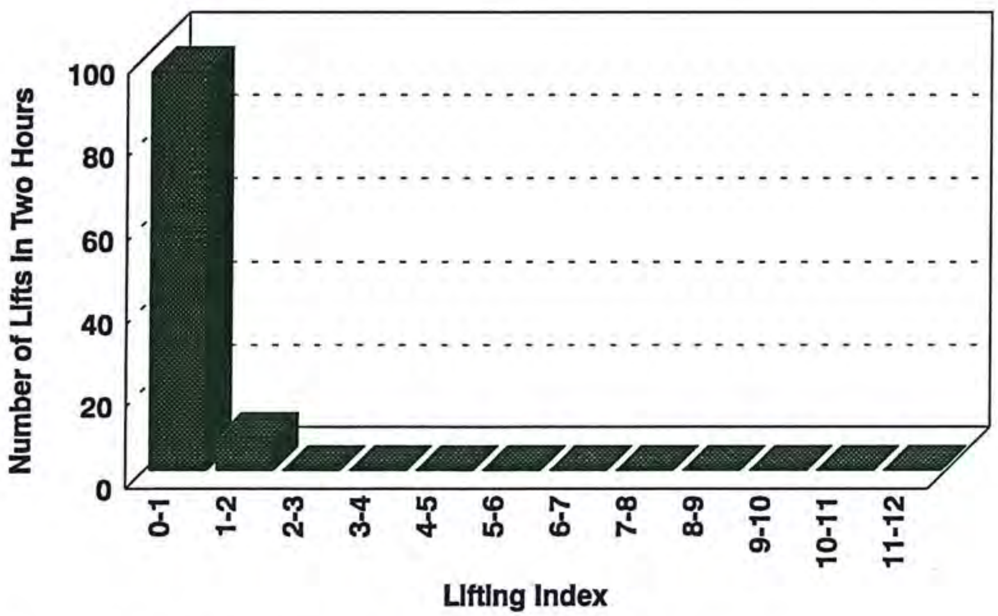

Figure 3. NIOSH Lifting Guide Assessment of a Mason

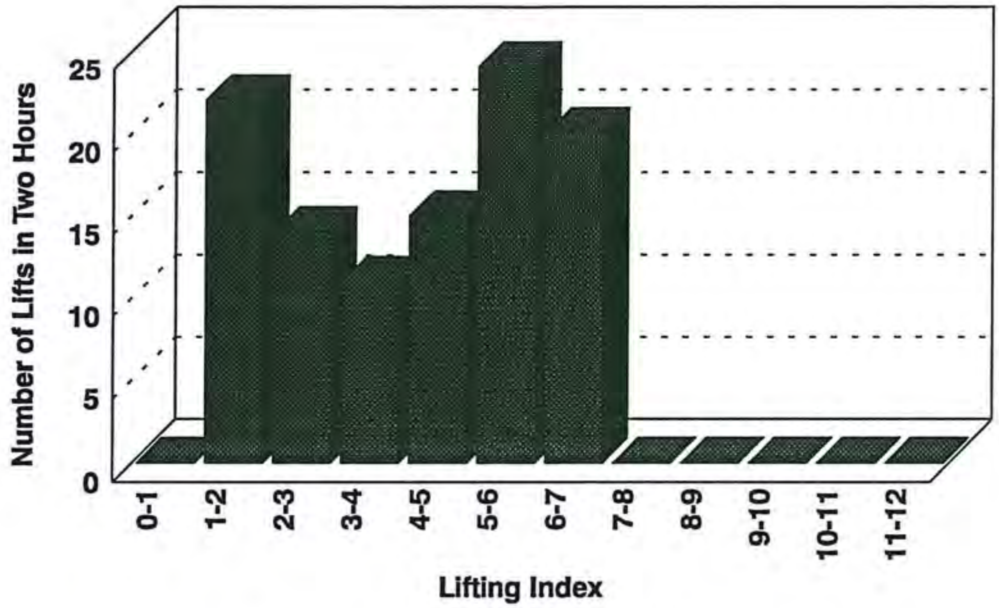

Figure 4. NIOSH Lifting Guide Assessment of a Mason's Helper

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