Proceedings of the Human Factors and Ergonomics Society Annual Meeting
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Proceedings of the Human Factors and Ergonomics Society Annual Meeting 1994 38: 578
DOI: 10.1177/154193129403801008

AN INVESTIGATION OF THE VARIABILITY IN HUMAN PERFORMANCE DURING MANUAL MATERIAL HANDLING ACTIVITIES

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The goal of this study was to quantify the variability of the three-dimensional kinematic and kinetic parameters describing the motion of the torso during the performance of sagittally symmetric lifting tasks. Subjects performed eight repetitions of simple lifting tasks described by 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). The three-dimensional, time dependent position, velocity and acceleration of the lumbar spine were monitored using the Lumbar Motion Monitor. These measures were then input into a dynamic biomechanical model which calculated torque about the L5/S1 joint in the sagittal plane. The results of the kinematic analysis showed significant variability in the magnitude of the peak velocity and acceleration in the sagittal plane and also showed significant motion in the transverse and coronal planes. The kinetic analysis showed an increase in the variability of the peak dynamic torque with greater levels of load but no coupling effect.

INTRODUCTION

Forces exerted on the spine during manual material handling (MMH) tasks can be classified into one of two categories: external and internal forces. To understand spine loading during MMH tasks, it is important to understand all aspects of these two categories of forces. One of the facets which has yet to be fully understood is the stochastic nature of these forces. 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 implies that a more appropriate method of analyzing the biomechanical system is to use a stochastic modelling approach. This stochastic modelling approach can be employed in two different ways: modelling variability in human performance (external) and modelling variability in the biomechanical stresses (internal).

With regard to the variability of the biomechanical stresses, one recent study (Mirka and Marras, 1993) has shown that there is a significant amount of variability in the internal stresses on the spine during lifting motions. In this study, subjects performed controlled trunk extension exertions. As the subjects performed these simulated lifting exertions, the electromyographic activities of ten trunk muscles were collected. The results of this study showed that there was significant variance in the muscle coactivation patterns employed during simple bending motions. These data were further examined using an EMG-driven biomechanical model and it was shown that the peak forces in the spine were 10% greater than the mean in compression, 40% greater than the mean in anterior shear and 50% greater than the mean in lateral shear. These results could be traced to the variable activation of the trunk musculature, specifically the erector spinae muscles. It was further shown that this erector spinae variability affected, through complex coactivation patterns, the shearing forces experienced by the spine. This study illustrated the potential variability of the internal stressors of the spine.

Based on some recent industrial surveillance studies, a measure of the variability of human performance during MMH activities may also be important in describing the etiology of low back injuries. Herrin et al (1986) performed detailed biomechanical analyses of 55 industrial jobs which entailed a total of 2934 MMH tasks. These authors 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.

Another industrial study (Marras et al, 1993) used a motion monitoring device to capture the continuous kinematic parameters that describe three-dimensional trunk motion during MMH tasks. Their results showed that the peak kinematic variables (such as peak velocity in the coronal plane) are more predictive of the risk of injury than are the average kinematic variables (such as average velocity in the coronal plane) for a particular task. These two studies indicate that by focussing 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. The objectives of the present study were to quantify the variability of the kinematic and kinetic parameters describing human performance of sagittally symmetric lifting activities and to investigate the effects of different workplace variables on the magnitude of this variability.

METHOD

Subjects

Seven male college graduate and undergraduate students served as subjects in this experiment. None had a history of low back impairment/disorder. Some static anthropometry is shown below.

Table 1. Subject Anthropometry

	Mean	Std
Age (yrs)	25	2.98
Mass (kg)	82.9	6.31
Stature (cm)	179.0	7.64
Spine Length (cm)	57.4	1.92
Shoulder Height (cm)	150.2	6.99
Trunk Circumference (cm)	90.1	6.56

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 revised lifting guidelines (Putz-Anderson and Waters, 1991). All lifts were performed in sagittally symmetric postures.

Dependent variables. The dependent variables were the kinematic parameters describing the position and motion of the lumbar trunk in three-dimensional space. Subsequent calculations using a dynamic biomechanical model of the torso allowed for the calculation of another dependent measure: total torque in the sagittal plane about the L5/S1 joint.

Design. Each combination of independent variables was repeated 8 times (4 lifts per minute for two minutes) per subject. The presentation of coupling conditions was randomized within load levels while the load levels began at 4.5 kg and then increased in 4.5 kg increments.

Apparatus

The motion of the lumbar trunk was monitored using the Lumbar Motion Monitor (LMM). This device was secured to the subject's back and measured the angular position, velocity and acceleration of the lumbar spine in the sagittal, coronal and transverse planes. The trunk position data was collected at a rate of 60Hz. For a more complete description of the LMM, see Marras et al (1992).

Subjects were asked to lift a 35x35x30cm 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. In the fair coupling condition, the subjects were instructed to lift the box from underneath, allowing the fingers to be flexed at 90°. In the fair condition, the box was placed on a stool to control for the distance the subjects had to stoop in order to reach the box. 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.

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 LMM would not inhibit normal trunk motion. Baseline trunk position readings were collected as the subjects stood in a relaxed vertical posture (0°) and in a sagittally symmetric 90° forward bend posture. These calibration data values would later be used in normalization of the data during 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 set using the randomization scheme. The lift rate was set at 4 lifts per minute, and the duration was two minutes for each experimental condition, thus rendering the eight repetitions. 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 constant degree of flexion throughout the range of lifting motion. Most subjects chose to lift with straight arms and a slight angle of flexion of 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.

Biomechanical Model

The biomechanical model calculated the instantaneous dynamic external torque 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. Using regression equations from Dempster (1955), and Pheasant (1986) as well as the static anthropometry of the subject population, estimates were made 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 collected at the beginning of the experiment. This normalized LMM data and the anthropometric values were then used as inputs to the dynamic biomechanical model which output the time dependent external torque in the sagittal plane about the L5/S1 intervertebral joint. The equation used to calculate the sagittal moment about L5/S1 is shown below:

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

<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 (kg)
R2 = distance from L5/S1 to shoulder joint (m)
α = angular acceleration of trunk (rad/sec/sec)
I = mass moment of inertia of the trunk (kg m^2)

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.

Data Processing

The first step in data processing was to normalize the data so that each trial began when the subject's hands came in contact with the box. In each plane of the body, the range of motion, peak velocity and peak acceleration were then obtained for the remainder of the lifting motion. Finally, the kinematic data from the LMM were input into the dynamic biomechanical model so that the time dependent torque about the lumbosacral joint could be plotted.

RESULTS AND DISCUSSION

Looking at the data qualitatively, Figure 1 illustrates the time dependent response of the acceleration in the sagittal plane. In this figure, time t=0 corresponds that point during the lift when the subject's hands first touched the box. This figure shows graphically the variability in the magnitude of the peak acceleration as well as the variability in the time history of acceleration.

The results of a more quantitative analysis of trunk kinematics showed significant amount of motion in all three planes. Comparing the three levels of kinematic data, the results of this study 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 10.4%.

From an ergonomic point of view the greatest impact of this 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 2. In an attempt to understand the effect of workplace factors on the magnitude of this variability, differences in the variance of the peak torques were tested using a Bartlett test. These results showed a significant effect due to the load weight, with greater weight levels showing a higher level of variability. This significant weight effect is graphically illustrated in Figure 2. Further analysis showed that coupling by itself had no significant effect on the magnitude of the variance, but at near maximal levels of load, poor coupling generated the greatest variability. Numerically, consideration of the range of potential torques about L5/S1 shows that the

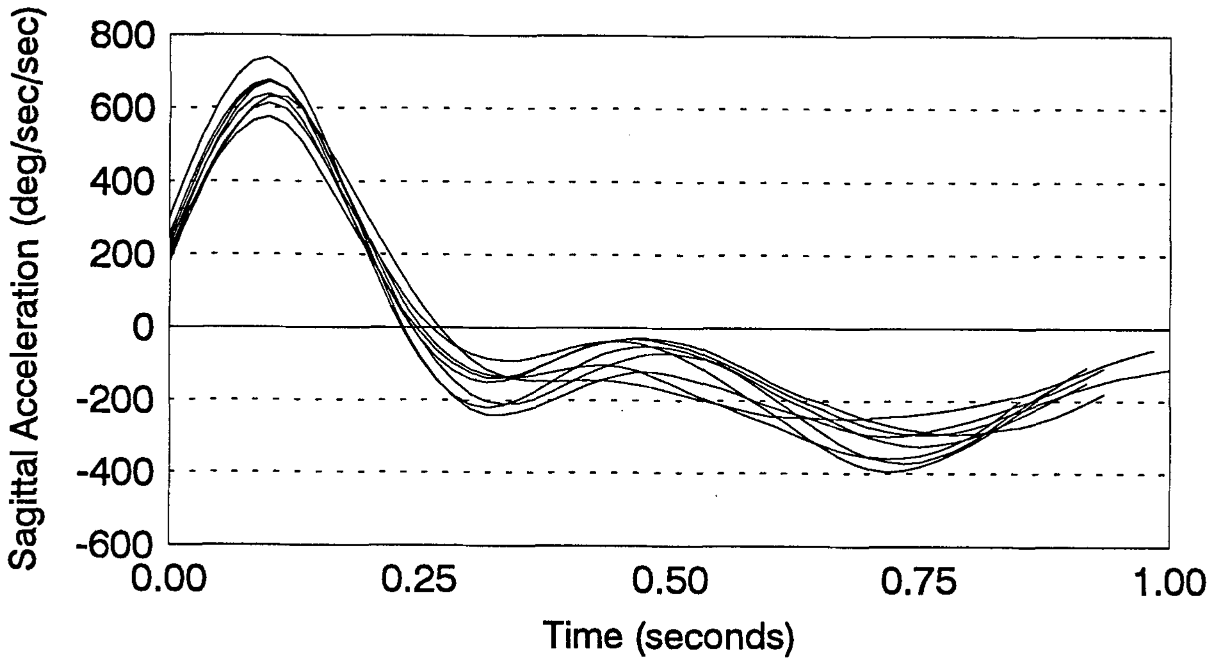

Figure 1. Time dependent sagittal acceleration. Load=22.5 kg, Coupling='Good'

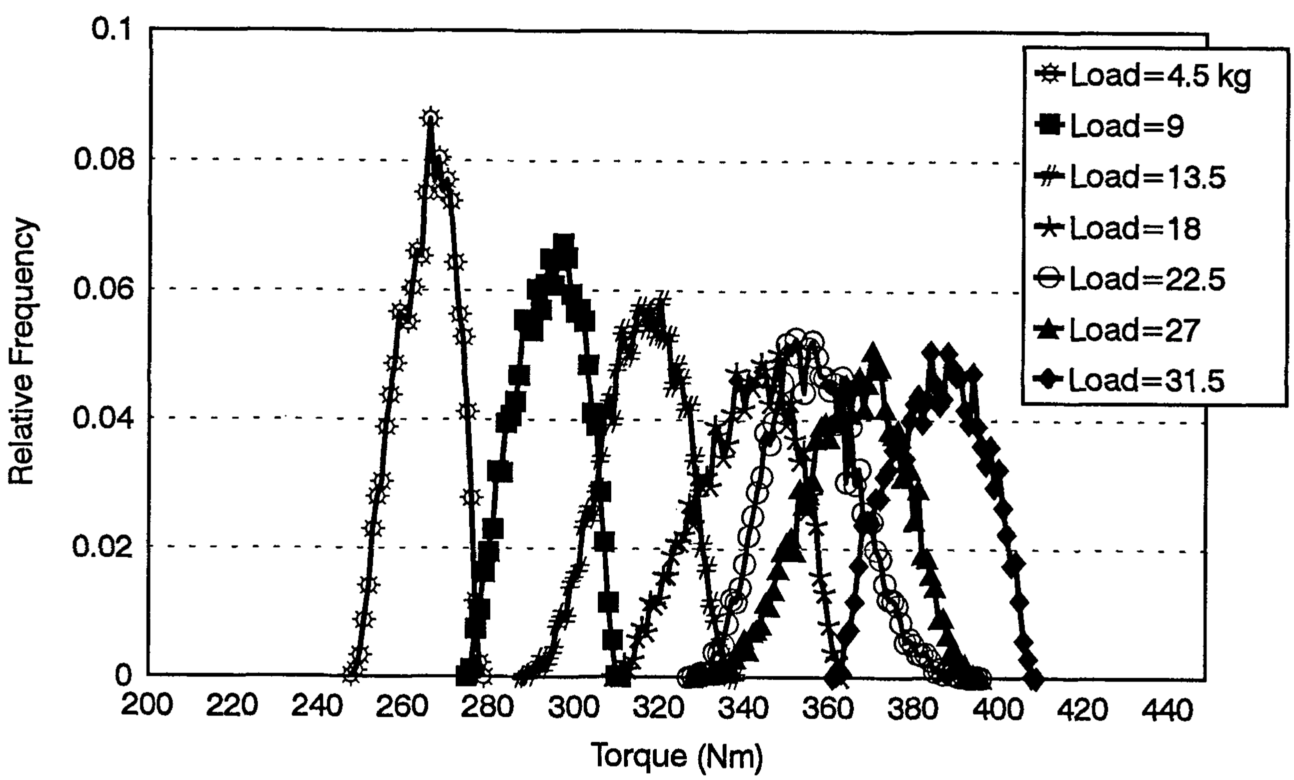

Figure 2. Distributions of peak sagittal torque as a function of load magnitude.

torques at 2 standard deviations above the mean are between 5% and 15% higher than those at the mean.

Table 2. Mean and standard deviation of torque in the sagittal plane (in Nm)

Coupling Quality

	Good		Fair		Poor	
	Mean	Std	Mean	Std	Mean	Std
Load						
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		

Kinematic results from the transverse and coronal planes showed that there was also a significant amount of dynamic activity in these "off-planes" even though the lifts were performed in sagittally symmetric postures. These results are shown in Table 3. These kinematic parameters did not, however, show significant trends as a function of load or quality of coupling.

Table 3. Mean and standard deviation of kinematic parameters in the coronal and transverse planes

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

CONCLUSIONS

This study has shown that there is significant variability in human performance- from both a kinematic and kinetic perspective. As ergonomists we need to be concerned with, or at least be aware of, the worst case scenario. This study has shown that by considering only the average performance profile, the peak torque is underestimated by between 5% and 15% and points to a need for the stochastic modelling of biomechanical systems to be considered.

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ACKNOWLEDGMENT

This publication was partially supported by grant number KO1 OH00135-01 from NIOSH. Its contents are solely the responsibility of the authors and do not necessarily reflect the official views of NIOSH.