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An investigation of the variability in human performance during sagittally symmetric lifting tasks

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Several industrial surveillance studies have suggested that variability in the biomechanical requirements of a job could hold significant information with regard to the risk of an individual incurring an occupational low-back injury. This variability may be in the form of variable job demands or in the form of variability in the technique that an individual uses to perform a task. The goal of the current study was to quantify the variability in human performance during sagittally symmetric lifting tasks. In this study, measures of human performance were limited to the three-dimensional kinematics of the lumbar region of the torso and the torque developed about the lumbosacral joint. Subjects were asked to lift a weighted box at a rate of four 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 they 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 with the Lumbar Motion Monitor. These postural and dynamic measures were then used to calculate the peak torque about the lumbosacral 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, even though these were sagittally symmetric lifting tasks. The kinetic analysis showed an increase in the variability of the magnitude of the peak dynamic torque with greater levels of load, whereas coupling showed little effect on the magnitude of this variability.

1. Introduction

Forces exerted on the spine during manual materials handling (MMH) tasks can be classified into one of two categories: external and internal. The external forces are generated outside 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 exerted by the musculoskeletal system. These internal forces include trunk muscle forces that provide the moments needed during lifting, as well as the compression, shear and torsional forces that 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 that has yet to be fully understood is its stochastic nature. The multidimensional, indeterminate nature of most biomechanical systems indicates that there are many ways in which a person can perform a movement or exertion – from both an internal and an external perspective. This suggests that a more appropriate method of analyzing the biomechanical system is to use a stochastic modeling approach that 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 (1993) 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 that precisely controlled the kinematics of the lifting motion. As the subjects performed these simulated lifting exertions, the electromyographic (EMG) activities of 10 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 10 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 torque exerted and the angular velocity of the lifting motion. These data were further examined by using an EMG-driven biomechanical model (Marras and Sommerich, 1991) 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, whereas 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.

On the basis of some recent industrial surveillance studies, a measure of the variability of human performance during MMH activities could also contain valuable information towards the description of the etiology of low back injuries. Herrin *et al.* (1986) performed detailed biomechanical analyses of 55 industrial jobs that 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 that 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 MMH tasks, valuable insight into workplace risk can be gained.

The objectives of the present study are threefold: first, to quantify the variability of kinematic parameters describing human performance of simple sagittally symmetric MMH activities; secondly, to describe how these variable kinematic data translate into variable torque about the lumbosacral (L5/S1) joint in the sagittal plane; and finally, to investigate the effect of workplace variables on the magnitude of this variability.

2. Method

2.1. *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 (s.d. = 2.98 years), a mean height of 179.0 cm (s.d. = 7.6 cm), and a mean mass of 82.9 kg (s.d. = 6.3 kg).

2.2. *Experimental design*

2.2.1. *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 (Waters *et al.*, 1994). All lifts were performed in sagittally symmetric postures.

2.2.2. *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 with a dynamic biomechanical model of the torso allowed for the calculation of a tenth dependent measure: torque in the sagittal plane about the L5/S1 joint.

2.2.3. *Design*

Each combination of independent variables was repeated eight times per subject. The presentation of coupling conditions was randomized within weight levels, whereas 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.

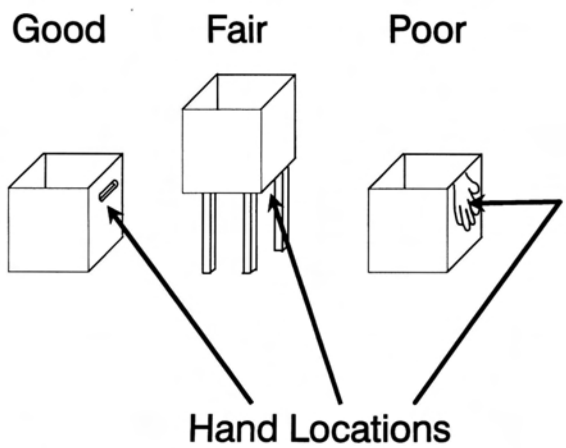

Fig. 1. Illustration of the hand placement locations in each of the three coupling conditions: good, use handles; fair, lift box from underneath; poor, compression lift on sides of box.

2.3. *Apparatus*

The motion of the lumbar region of the trunk was monitored by 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 obtain angular velocity and angular acceleration in the three cardinal planes. The three-dimensional trunk position data were collected at a rate of 60 Hz. For a more complete description of the LMM and the processing of the data, see Marras *et al.* (1992).

Subjects were asked to lift a 35 cm × 35 cm × 30 cm 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 Fig. 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°. In the poor coupling condition, the subjects were instructed to 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.

2.4. *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 three-dimensional 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 four lifts per minute and the duration was three minutes for each experimental condition. The data from the first minute were discarded to allow 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.

2.5. *Data processing*

The results of the data collection revealed that the trials in which 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 of a possible 56 (7 subjects; 8 repetitions per subject)).

The first step in data processing was to standardize the data by eliminating those data points that 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 used as input in a dynamic biomechanical model so that the peak torque about the lumbosacral joint could be calculated.

The biomechanical model calculated the time-dependent dynamic external torque about the lumbosacral joint. The body was partitioned into a five-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 anthropometric data of the each of the subjects were combined with data and regression equations from Chandler *et al.* (1975), Dempster (1955), Pheasant (1986) and Reynolds (1978) to obtain estimates of (1) the trunk mass, (2) the

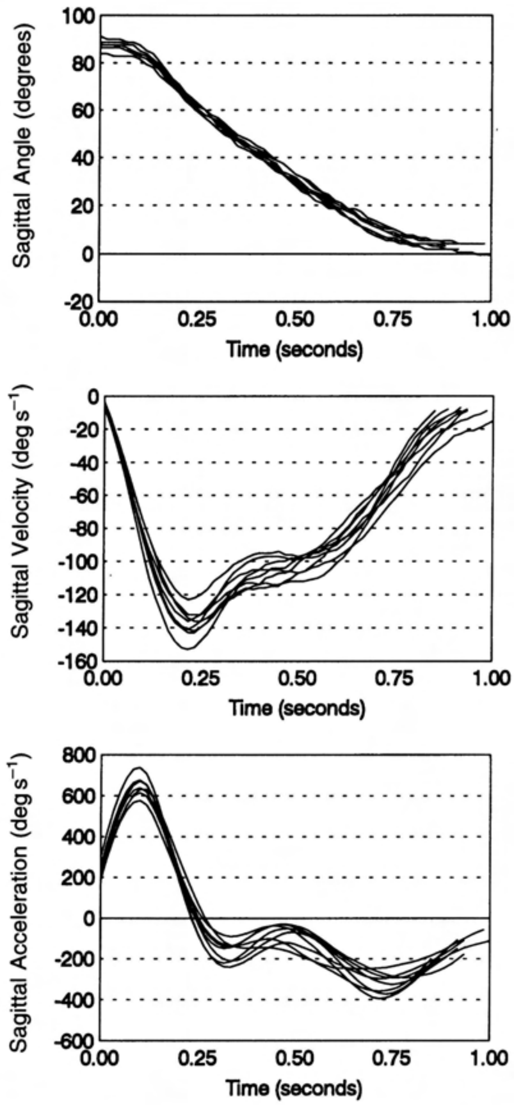

Fig. 2. Time-dependent traces of sagittal angle, sagittal velocity and sagittal acceleration: load = 22.5 kg; coupling = Good.

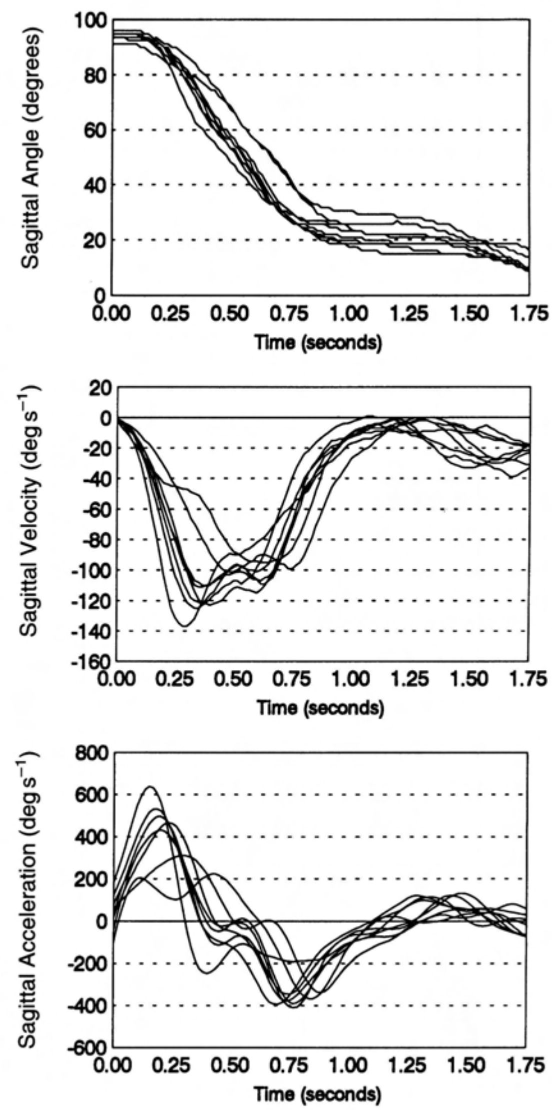

Fig. 3. Time-dependent traces of sagittal angle, sagittal velocity and sagittal acceleration: load = 22.5 kg; coupling = Poor.

distance between the center of mass of the 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 were 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 torque in the sagittal plane about the L5/S1 intervertebral joint. The equation used to calculate the sagittal moment about L5/S1 is

$$M(\mathrm{L5/S1}) = M_{\mathrm{T}} g R_1 \sin\theta + M_{\mathrm{AB}} g R_2 \sin\theta + M_{\mathrm{AB}} R_2^2 \alpha \sin\theta + I\alpha, \qquad (1)$$

where

M_{T} = mass of trunk (kg);
g = gravitational constant (9.81 m s^{-2});
R_1 = distance from L5/S1 to center of mass of trunk (m);
θ = sagittal angle of the trunk (upright = 0°);
M_{AB} = mass of the arms plus mass of box and contents (kg);
R_2 = distance from L5/S1 to gleno-humeral joint (m);
α = angular acceleration of trunk in the sagittal plane (rad s^{-2}); and
I = mass moment of inertia of the trunk and head about L5/S1 (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 60 data points per second.

Note that the third term in (1) 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. To verify that this is an appropriate assumption for the current experiment, subjects stood on a force platform during lifting. The force platform data were used to verify that the peak vertical ground reaction forces (and thereby the peak torque values) occurred before significant horizontal ground reaction forces were developed.

3. Results

Looking at the data qualitatively, Figs 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 means and standard deviations 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 were 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 whether 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 (c.v.) for the range of motion in the sagittal plane was 2.5% whereas the c.v. for peak velocity was 8.1% and the c.v. 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.

Table 1. Means and standard deviations of peak kinematic parameters in the sagittal plane

Load (kg)		*Quality of hand–container coupling*					
		Good		*Fair*		*Poor*	
		Mean	*s.d.*	*Mean*	*s.d.*	*Mean*	*s.d.*
4.5	ANG (deg)	87.6	1.8	90.8	1.3	92.7	1.1
	VEL (deg s^{-1})	−165.2	11.4	−172.5	8.6	−166.6	9.5
	ACC (deg s^{-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 s^{-1})	−166.6	12.9	−162.5	9.6	−166.6	10.0
	ACC (deg s^{-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 s^{-1})	−158.9	11.9	−155.8	10.1	−154.3	10.3
	ACC (deg s^{-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 s^{-1})	−153.0	10.0	−149.9	11.2	−137.1	10.6
	ACC (deg s^{-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 s^{-1})	−137.5	13.0	−137.3	8.2	−119.3	16.1
	ACC (deg s^{-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 s^{-1})	−131.8	13.8	−122.9	8.8		
	ACC (deg s^{-2})	563.7	83.0	521.9	53.7		
31.5	ANG (deg)	85.2	2.6	86.1	1.6		
	VEL (deg s^{-1})	−124.3	12.4	−121.1	8.0		
	ACC (deg s^{-2})	504.7	56.4	485.5	43.7		

Table 2. Means and standard deviations of the peak kinematic parameters in the coronal and transverse planes

	Mean	*s.d.*
Coronal range of motion (deg)	4.6	1.7
Transverse range of motion (deg)	3.1	1.7
Max. coronal velocity (deg s^{-1})	13.6	4.6
Max. transverse velocity (deg s^{-1})	7.8	3.1
Max. coronal acceleration (deg s^{-2})	63.2	22.6
Max. transverse acceleration (deg s^{-2})	40.4	13.7

Table 3. Means and standard deviations of peak torque (N m) in the sagittal plane

Load (kg)	*Quality of hand to container coupling*					
	Good		*Fair*		*Poor*	
	Mean	*s.d.*	*Mean*	*s.d.*	*Mean*	*s.d.*
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		

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

In an attempt to understand the effect of workplace factors on the magnitude of this variability, differences in the variances of the peak torques were tested with 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 were from a normal distribution. The lowest W found was 0.91 and the average across cells was 0.96, indicating that these data fit the normal distribution. Due to the unbalanced nature of the data set, the data set analyzed by 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 were analyzed. These results showed a significant effect ($B = 47.75$; $\chi^2_{(0.05,6)} = 12.592$) of the load weight, with greater weight levels showing a higher level of variability. This significant weight effect is illustrated graphically in Figure 4. (The distributions shown in this figure are parametrized Johnson distributions that were fit to the empirical data of this experiment. See DeBrota *et al.* (1989) and Mirka and Marras (1993) for further details on fitting empirical data to the Johnson family of distributions.) To test for a coupling main effect, only the load levels that were less than 27 kg were considered. The Bartlett test for equality of variance revealed that coupling as a main effect had no significant effect on the magnitude of the variance.

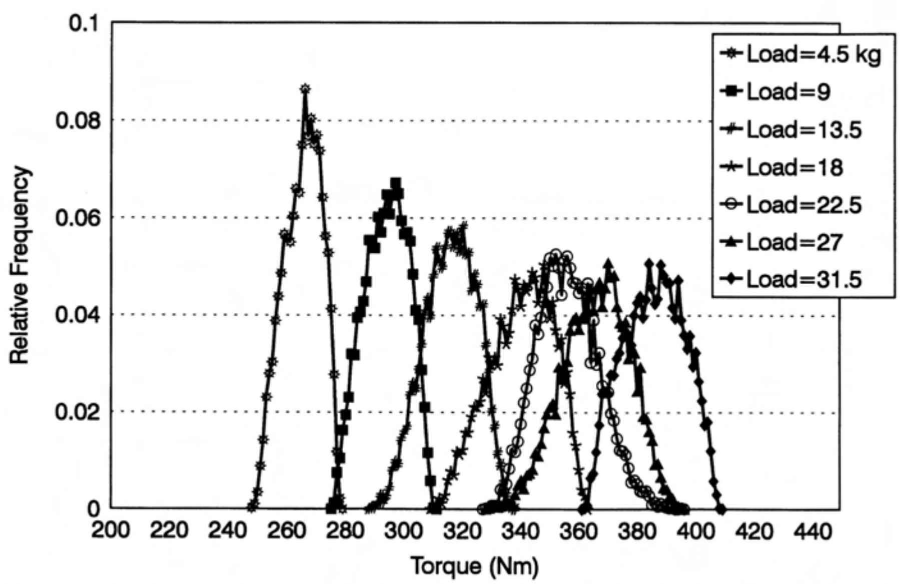

Fig. 4. Distributions of peak sagittal torque as a function of load magnitude. Note that in addition to the increase in the mean at higher load levels, there is also an increase in the variability.

An analysis of the response of the means showed that load level had a significant ($P < 0.0001$) effect on the magnitude of the peak torque, whereas coupling effects were shown to be significant only under specific load levels. To make the data collected in this study useful to individuals wishing to use them in a stochastic model of human performance, regression analysis was performed on the peak sagittal torque data. Variables considered for this equation included: subject weight, subject height, load and (load)^2. The following equations were the best fit to the empirical data

$$\text{mean (N m)} = 248.47 + 5.023 \text{ load (kg)} - 0.0220 \text{ (load (kg))}^2 \quad (R^2 = 0.874) \quad (2)$$

$$\text{s.d. (N m)} = 1.16 \text{ load (kg)} - 0.0238 \text{ (load (kg))}^2 \quad (R^2 = 0.961) \quad (3)$$

4. Discussion

Variability in the biomechanical aspects of human performance is a research topic that has received little 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 can now be affected by the worker's lifting strategy. 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, 1990, 1992, 1993). 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 that has shown how variability is affected by extreme lifting conditions. Previous work done by Parnianpour *et al.* (1988) 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 nonfatigued conditions – a result that indicates that current research may be oversimplifying 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 (Shirazi-Adl, 1984).

When the human performance variability demonstrated in this study is combined with the variability in internal forces shown by Mirka and Marras (1993), it becomes evident that 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 torque is underestimated by between 5% and 11%, for 1 lift out of every 50. This was the result in an experimental environment where many factors found on a plant floor that could 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 distractions that 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 from these 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 were 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 the risk of low-back injury could be the configuration that 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 might begin to concern themselves with variance reduction/control procedures when designing workspaces to reduce musculoskeletal injuries.

5. 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 that may be better able to predict risk of injury for occupational manual material handling tasks.

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