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A Three-Dimensional Motion Model of Loads on the Lumbar Spine: II. Model Validation

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A three-dimensional motion model has been developed that estimates loads on the lumbar spine under laboratory conditions that simulate manual materials handling conditions. Eleven subjects experienced spinal loading during an experiment in which conditions of trunk velocity, trunk torque output, and trunk asymmetric posture were varied in a series of isokinetic velocity trunk extensions. The electromyographic activity of 10 trunk muscles, subject anthropometry, and trunk kinetics were used as input to a biomechanical simulation model described in Part I of this study. The model calculated estimates of compression, shear, and torsion loading in the lumbar spine, as well as the torque production of the trunk, continuously throughout the exertion. Trunk torque estimates derived from this model were compared with measured trunk torque. The effects of trunk motion, posture, and torque level on spine loading as estimated by the model are discussed. It was concluded that this approach provides a straightforward means of assessing loading of the spine attributable to laboratory simulations of workplace conditions.

INTRODUCTION

Despite increasing automation in the workplace, incidents of low back disorders (LBD) continue to plague humankind. Both epidemiological (Andersson, 1981) and biomechanical (Chaffin and Park, 1973) studies have indicated that there is a link between the risk of LBD and occupational requirements. Specifically, lifting and other forms of manual materials handling (MMH) are associated with greater risk of LBD.

Most biomechanical efforts aimed at reducing the risk of occupationally related LBD have focused on the activities of the trunk musculature. These muscles provide a restorative moment during MMH. The forces generated by these muscles could easily become excessive given that their moment arms relative to the spine are short compared with the moment arm between the spine and the object being lifted. Thus it is extremely important to understand how the trunk muscles collectively load the spine under occupational conditions.

Many researchers have attempted to simulate workplace conditions, such as those encountered in manual materials handling, in the laboratory in order to assess occupational

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trunk loading. Originally these assessments were confined to sagittal plane, isometric trunk exertions. Two-dimensional, static models were traditionally used to represent these conditions and to estimate the compressive forces acting on the spine. Most of these early models based their estimates of spine loading primarily on the external moments imposed around the trunk. If internal contributions to spine loading were considered, they were usually limited to estimates of trunk muscle forces represented by a single equivalent posterior trunk muscle.

As technology advanced, researchers discovered means of measuring and controlling the trunk under three-dimensional dynamic motion conditions. As mentioned in the companion article (Part I) to this study (Marras and Sommerich, 1991 [this issue]), these studies estimated that spine loading increased between 22.5% and 60.0% under dynamic conditions compared with static conditions. These studies utilized kinetic information about body motion to estimate moments of inertia imposed around the body links. Some of these models do include internal muscle force estimates in the evaluation of spine compression, but none includes the effects of muscle coactivation. Furthermore, these models usually do not use empirical measures to validate the models or customize the results to particular individuals.

In the companion article (Part I) to this paper, we described a model that uses muscle electromyography of 10 trunk muscles, subject anthropometry, and kinetic information about the back motion to estimate spine compression, shear, and torsion and trunk torque. This model is intended to be a research tool to investigate the effects of constant velocity trunk motion and trunk asymmetric position on the loading and torque production capabilities of the trunk. The advantage of this model is that it is capable of estimating trunk loading continuously throughout the exertion. The model also includes the effects of muscle coactivity throughout time, as well as effects of asymmetry and motion. Additionally, this model can be tailored to an individual because it considers anthropometry and the specific individual muscle actions of the subject.

The primary objective of this study was to test the validity of the model experimentally. Unfortunately, it was not possible to directly measure the compression, shear, and torsional loading forces acting on the spine in vivo. However, according to the force and moment equations that govern trunk loading (Schultz and Andersson, 1981), trunk torque is directly related to spine loading. Therefore, if we assume that these equation relationships are correct, we could validate the model by comparing measured trunk torque production with trunk torque production estimated by the model. Thus the goal of this paper was to compare and describe the performance of this model using trunk torque as an indicator of performance. In this study we were interested in the performance of the model in response to changes in the trunk activity parameters usually dictated by the workplace. These parameters include trunk velocity, trunk asymmetry, and load level. An additional goal of this paper was to describe how spine loading changed as these variables were systematically altered.

METHOD

Approach

In order to achieve the objective of this study, it was necessary to preserve the relationship between muscle force and electromyographic activity. A highly controlled experiment was designed wherein subjects were required to exert a constant torque with the back throughout a 45-deg range of motion while moving at constant velocities. Subjects extended the trunk in a simulated lifting motion under both sagittally symmetric and asymmetric trunk positions. It was assumed that the point of bend around the spine was located at L5/S1 and that this motion would relate to torques experienced around the spine during lifting.

Subjects

Eleven male volunteers served as subjects in this experiment. None of the subjects had experienced a significant low back disorder, and all were considered to be in good health. Subject occupations covered a wide range, from professionals to students to those experienced in manual materials handling (MMH). The anthropometric characteristics of the subject population are presented in Table 1.

Design

The independent variables consisted of three trunk position-loading combinations and three different angular trunk velocities. Trunk position-loading combinations consisted of (1) a sagittally symmetric trunk extension while 27.1 Nm of torque were generated with the back, (2) a sagittally symmetric trunk extension while 54.2 Nm of torque were generated with the back, and (3) an asymmetric trunk exertion involving a 30-deg twist away from the sagittal plane while generating 27.1 Nm of torque with the back. In the asymmetric position the subject's trunk was rotated clockwise (30 deg) with respect to the pelvis.

Each of these position-loading combinations was tested under three isokinetic, angular velocity conditions consisting of 10, 20, and 30 deg/s. The levels were chosen to fall within MMH back velocity ranges reported by Kim and Marras (1987).

In this experiment the electromyographic activities of the 10 trunk muscles used by the model were sampled. These trunk muscles consist of latissimus dorsi right (LATR) and left (LATL), the erector spinae right (ERSR) and left (ERSL), the external oblique right

TABLE 1

Subject Anthropometry

Subject	*Age (yrs)*	*WT (kg)*	*HT (cm)*	*TR (cm)*	*B (cm)*	*D (cm)*
1	25	72.7	177.8	63.5	33.0	22.9
2	25	72.7	180.2	61.9	31.1	21.4
3	34	70.5	171.5	56.8	30.8	23.4
4	28	69.5	172.6	57.0	31.2	22.3
5	21	102.3	191.5	65.1	24.5	34.7
6	34	84.1	184.3	65.0	22.5	30.6
7	23	79.5	185.1	62.2	31.2	21.5
8	22	65.9	179.2	55.3	28.0	17.6
9	22	79.1	182.3	59.6	29.9	21.0
10	22	73.6	175.6	53.7	27.7	19.9
11	22	70.0	177.2	59.5	29.2	20.2
Avg.	25.3	76.4	179.8	60.0	29.0	23.2
S.D.	4.8	10.1	5.8	3.9	3.2	5.0
C.V.	18.8	13.2	3.3	6.5	10.9	21.5

WT = total body weight, HT = total height, TR = trunk length, B = trunk breadth at L5, D = trunk depth at L5.
Avg. = mean, S.D. = standard deviation, C.V. = coefficient of variation = (S.D./Avg.) × 100.

(EXOR) and left (EXOL), the internal oblique right (INOR) and left (INOL), and the rectus abdominus right (RCAR) and left (RCAL). Lippold (1952) and Bigland and Lippold (1954) have shown that the integrated EMG activities of a muscle operating under isometric or constant velocity conditions are related to the amount of force the muscle is producing.

Apparatus

The configuration of the equipment used in this experiment is shown in Figure 1. Velocity was controlled by a KIN/COM isokinetic dynamometer. This device was aligned with the L5/S1 junction of the back via an asymmetric reference frame (ARF). This ARF positioned the subject relative to the dynamometer so that both symmetric and asymmetric back exertions could be tested.

Trunk torque around L5/S1 was controlled by the subject. The subjects viewed a computer monitor that graphically displayed their current level of torque production on line. A target torque was displayed on the computer screen, as was a tolerance band of ±10% around the target torque. Subjects were able to monitor their torque production continuously and to use this feedback to maintain the specified torque level. EMG activities of the trunk muscles were monitored using small surface electrodes attached to small, lightweight preamplifiers. These preamplifiers were mounted on a belt that was secured around the subject's waist. This configuration minimized the amount of noise in the recorded signal. The preamplifiers were connected to EMG amplifiers, filters, and integrators. A switchbox was connected to the EMG amplifiers, which permitted the signal quality of each EMG signal to be monitored. The EMG signal was low-pass filtered at 1000 Hz and high-pass filtered at 80 Hz. The signal was rectified and averaged with a time-constant window of 20 ms. This served as the integrated signal.

The dynamometer signals, ARF position, and EMG signals were all digitized with an analog-to-digital (A/D) converter. This multichannel A/D system interfaced with a 386-based microcomputer to collect, display, and store the data on-line.

Procedure

Subjects were interviewed to ensure that they had not experienced any significant back

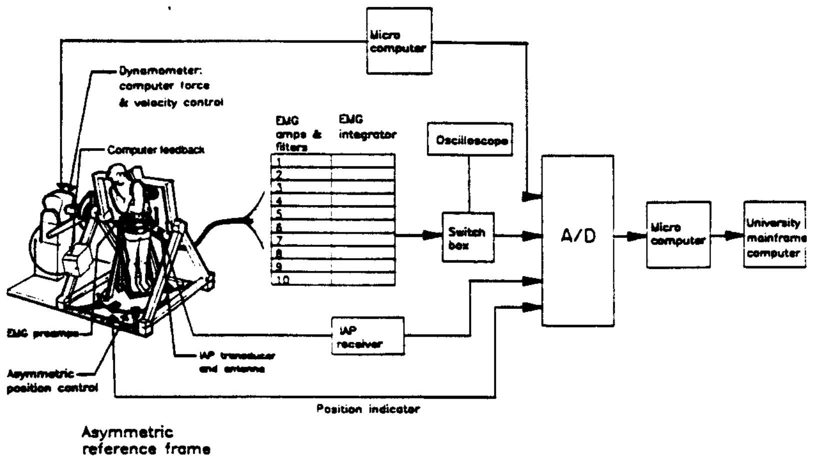

Figure 1. *Test equipment used in experimental paradigm.*

disorders. A practice session was permitted in order to familiarize each subject with the experimental task of controlling torque by interfacing with the computer feedback system. Once a subject became proficient at the task, an appointment was made for the experimental session, which was always conducted on a subsequent day. During the experimental session the subjects were connected to the EMG system via standard preparation procedures. After all signal qualities were verified, the experiment began. Maximum and minimum activities of each muscle were collected with the trunk in forward bending positions of 5.0, 22.5, and 40.0 deg in both the sagittally symmetric and asymmetric trunk positions. These EMG recordings were collected for normalization purposes as well as to determine the length-tension relationship for each muscle. A minimum of 2 min of rest was allowed between exertions.

The experimental task required subjects to control their trunk torque between the tolerance limits of the exertion (defined on the computer screen) under each condition. If the subject failed to maintain exertions within the tolerance limits, the trial was rerun.

Data Conditioning

The data were processed according to the procedures described in Part I of this paper (Marras and Sommerich, 1991). These procedures generally consist of identifying four event points for each EMG signal and processing the EMG inputs. The reader is directed to Marras and Sommerich (1991) for a complete description of the data-processing procedure.

RESULTS

The items to be discussed in this section involve both model parameters and model outputs, including the gain value and its relation to subject anthropometry, the effects of asymmetry and external load level on muscle output, the correlation between measured and calculated torque, and the effects of velocity, trunk symmetry, and load level on peak compression.

Gain

Gain is the term that encompasses muscle force per unit area. This value also contains the combined adjustments, or fine tuning, for the EMG modulations and for the calculated muscle areas. The gain, determined separately for each subject under each test condition by the model (as described in Part I of this study), was found to be strongly related to subject anthropometry. A regression equation (Equation 1) was developed from subject weight, trunk length, and depth and breadth measurements at L5 for the purpose of determining the average gain for each subject across loading conditions. This relationship is also depicted in Figure 2.

$$Gain = 36.64 \times WT - 46.12 \times TL + 0.521 \times WT \times TL + 8.49 \times B + 7.23 \times D + 2808.6, \quad (1)$$

where WT = subject weight, TL = trunk length (cm), B = trunk breadth at L5, and D = trunk depth at L5.

The R^2 associated with this regression line is 0.75. This is interpreted as a 75% reduction in the variation associated with predicting the average gain for a subject when using the selected anthropometric information. As a point of interest, the R^2 can be improved to 0.98 by using nine of the eleven subjects' data. The two subjects who were removed may not, for reasons not fully understood at this point, be compatible with assumptions made within the model. There is reason to believe that one subject in particular may have been operating under his own experimental hypothesis, which may have influ-

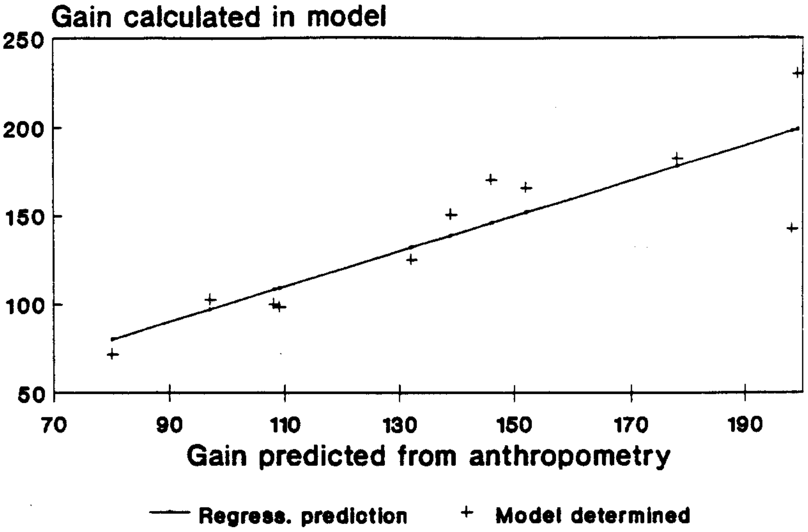

Figure 2. *Average gain per subject under all loading conditions. Actual values are presented as crosses. The regression line, calculated from anthropometric data, accounts for 75% of the variation in predicting these values.*

enced his performance. In a large group of subjects one subject's data may not have the impact that it could have in a smaller study, such as the present one.

Muscles

The effects of trunk symmetry and external load level on each of the 10 trunk muscles are depicted in Figure 3. The effects are readily apparent in the erector spinae, the major trunk extensors. For symmetric trunk orientation the average forces are almost equal for the left and right pair. However, under the asymmetric condition the left muscle is dominant. Elevated muscle force, associated with an increase in external loading, is seen for the erector spinae and internal oblique muscles.

Torque

For each trial the lateral torque (torque around the x axis) calculated by the model is compared with that measured during a subject's extension exertion trial for the purpose of determining model performance. The total area under each curve is compared, as well as individual values, when the calculations are performed within the model. Well over 85% of the torque pairs (measured and calculated) had R^2 values of 0.7 or greater. See Figure 4 for a pictorial representation of the frequency distribution of these correlations.

Calculated lateral torque varied as a function of trunk symmetry and external load level. These effects can be seen in Figure 5. The estimated average maximum lateral torque is greater for symmetric trunk orientation and for higher external load levels.

Compression

For the symmetric conditions increasing peak compression levels were calculated for both increases in external loading and increases in velocity (see Figure 6). This trend did not appear in the asymmetric conditions, where peak compression was approximately the same for each velocity condition (see Figure 7).

Peak anterior-posterior shear and peak

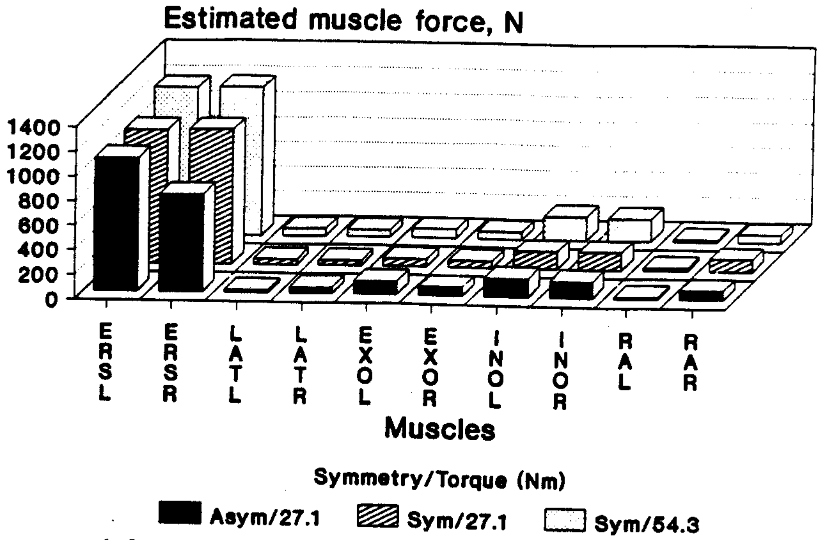

Figure 3. *Average muscle force, over all subjects, for each of the 10 muscles as a function of trunk symmetry and external load level.*

compression were both seen to increase with increases in symmetry and external load. Right-left shear was highly variable between subjects, so no trends could be discerned. The increase in right-left shear under the asymmetric conditions shown in Figure 7 does make sense from a biomechanical standpoint: one would expect that as the body moved more asymmetrically, the components of muscle force that would tend to resist lateral motion of the body would increase, thereby increasing right-left shear.

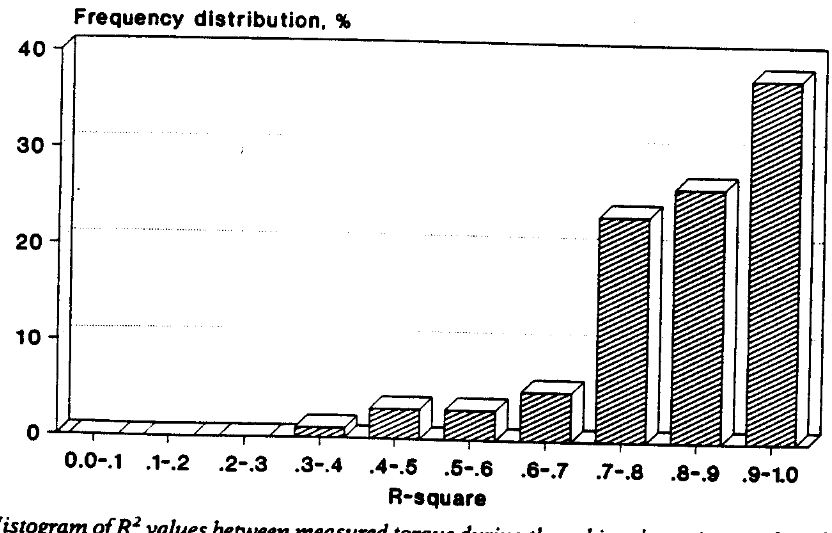

Figure 4. *Histogram of R^2 values between measured torque during the subjects' exertions and model-calculated torque (total of 98 trials).*

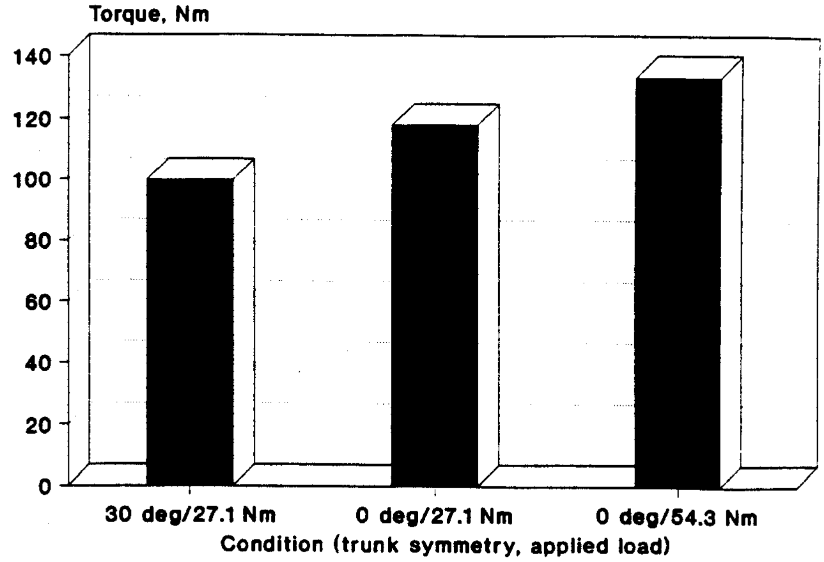

Figure 5. *Model-calculated peak torque for all subjects as a function of trunk symmetry and external load level.*

DISCUSSION

Generally this model performed well in validation testing. The model performance measure—torque—indicated that the model was able to predict trunk torque very well. Most model runs were capable of explaining more than 70% of the variability in trunk torque production. This is regarded as a good correlation, considering that the model was run under both motion and asymmetric loading conditions.

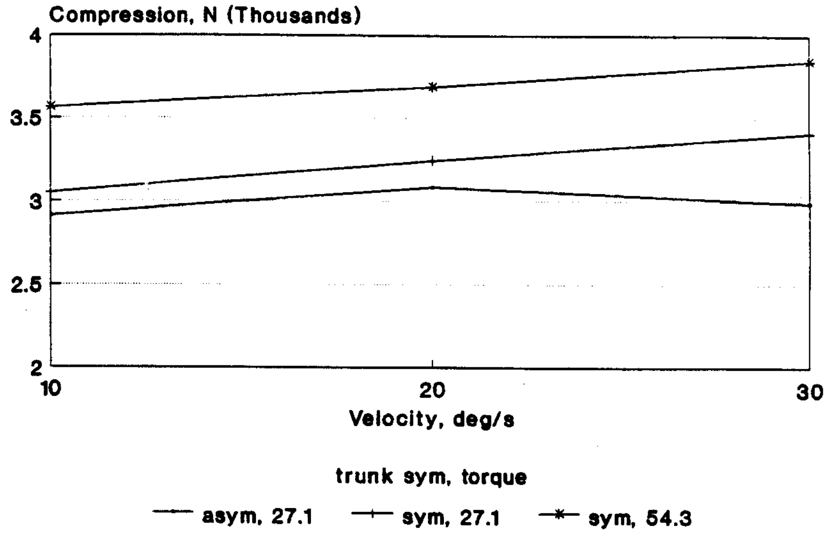

Figure 6. *Model-calculated peak spinal compression as a function of velocity, trunk symmetry, and external load level.*

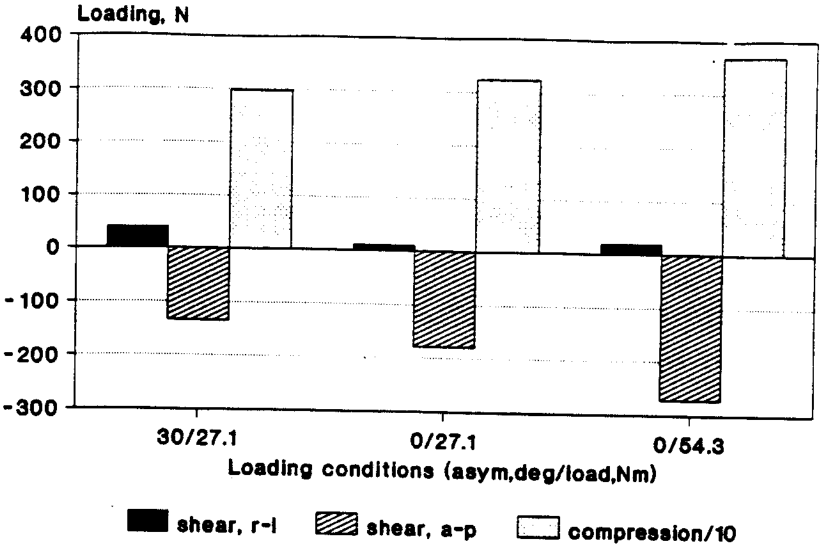

Figure 7. *Model-calculated average peak spinal loading as a function of trunk symmetry and external load level.*

Assuming that the trunk force and moment equations developed by Schultz and Andersson (1981) are correct, we are able to gain some insights as to the effects of trunk velocity and trunk asymmetry. This experiment was designed so that we could consider the effects of trunk torque, velocity, and asymmetry independently. As expected, the results indicated that as the trunk load or torque level increased, the compression of the spine increased. However, we were also able to discern trends related to velocity and asymmetry which were not obvious without the use of the model. Compared with sagittally symmetric lifts, mean spine compression was reduced for a given force level as the trunk became more asymmetric.

There was also a trade-off between spine compression and shear. Mean right-left lateral shear increased along with trunk asymmetry. This finding must be interpreted with caution, however, given that significant variance between subjects was noted for this parameter. A three-dimensional recording of measured trunk torques might have helped to explain this variance. Nonetheless, epidemiologic studies have shown that the risk of a low back disorder is greater when working in asymmetric or twisting positions (Andersson, 1981). This study suggests, however, that this increased risk may be a function of increased shear forces acting on the spine, not increases in compressive loads. This type of information should be considered in future lifting models and guides. Existing models and guides base their risk assessment solely on spine compression (National Institute for Occupational Safety and Health, 1981).

This model has also shown that spine compression in sagittally symmetric conditions increases by approximately 100 N for every 10 deg/s increase in trunk angular velocity. This trend may have profound effects on the risk associated with certain MMH tasks. For example, Figure 6 shows that during a sagittally symmetric exertion, where 27.1 Nm of trunk torque are produced at a velocity of 10 deg/s, about 3100 N of spine compression is present. This amount of loading would be considered safe (below the action limit) by current lifting guidelines (National Institute for Occupational Safety and Health, 1981). However, if this same exertion were performed at a velocity of 30 deg/s, spine compression would surpass the action limit (more than 3400 N), and one would expect that the task would begin to present a risk of suffering a vertebral end-plate microfracture. Thus it is important to understand the effects of trunk motion on spine loading. This information could also be of value in workplace and lifting design.

Several factors may either facilitate the model accuracy or account for differences between the model predictions and observed behavior. First, this was a three-dimensional model, but the external torque produced by the trunk was monitored in only one plane of motion. Thus any secondary torques exerted in the lateral or transverse planes were not detected by the instrumentation. Such conditions could be expected, especially in the asymmetric conditions. This situation would result in increased muscle activities and predictions of torques in the other planes of the body that were not measured. This, in turn, would affect the calculation of the gain factors in the model. Future studies should include three-dimensional measures of trunk torque so that these secondary torques could be measured. In this case even better correlations between predicted and measured model performance would be expected.

Second, the model could also be improved with more accurate estimates of muscle cross-sectional areas, as well as improved approximations of muscle locations relative to the spine. Both of these measures could be determined with magnetic resonance imaging (MRI) or computer tomography (CT scan) technology. In this manner the model could be custom-fit to particular subjects.

As mentioned earlier, this model is intended to be used to interpret laboratory data. However, the information gained in this study could also be used to gain insight into workplace design effects. Even though the motions studied in this experiment were performed at constant velocity, this information brings us one step closer to understanding how the spine is loaded under free dynamic working conditions. Findings such as the effects of asymmetry and trunk velocity could be extrapolated in a general sense to dynamic lifting conditions. Recommendations would include designing workplaces so that trunk motion requirements are minimized under various asymmetric motion conditions.

Further model enhancements could be considered at this time. These might include additional dynamic components. The model might be adjusted to consider trunk acceleration by including an acceleration modulation factor. Another improvement to the model might be the incorporation of additional muscles. For example, the psoas muscles could be incorporated if one could devise a means to record their electromyographic activity. Additionally, future versions of this model may incorporate free dynamic motion of the trunk. Before that could be realized, however, a clearer understanding of the relationship between electromyography and muscle force would be necessary. Finally, this model could be used in conjunction with whole-body biomechanical models. A whole-body model could determine the external forces or torques imposed on the body during a manual materials handling task. The current model would work in concert with these torque predictions to estimate the associated spine loading forces.

SUMMARY

The objectives of this paper were to investigate how well this motion model simulated the action of the trunk and to investigate loading changes of the spine as the trunk performed lifting tasks under dynamic, asymmetric conditions. Based on the measurable

performance parameters, the model appears to be quite robust and does reflect the action of the trunk under controlled velocity in symmetric and asymmetric conditions. The model indicated that spine compression increased directly with trunk velocity. It also indicated that as the trunk became more asymmetric, spine compression decreased but shear forces increased. We also concluded that better instrumentation of the experimental task and more accurate model input information, such as MRI or CT scan information, would enhance model performance.

ACKNOWLEDGMENTS

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