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PREDICTIONS OF FORCES ACTING UPON THE LUMBAR SPINE UNDER ISOMETRIC AND ISOKINETIC CONDITIONS: A MODEL–EXPERIMENT COMPARISON

W.S. Marras

Department of Industrial and Systems Engineering, The Ohio State University, Columbus, OH 43210 (U.S.A.)

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ABSTRACT

The biomechanical analysis technique of Schultz and Andersson (1981) was used to predict sagittally symmetric activity of the trunk under motion conditions. These biomechanical predictions were compared with the results of experimental exertions of 20 male and female subjects performing both isometric and isokinetic back extention activities. The results indicate differences in behavior of the internal trunk structures under motion conditions compared with quasi-static model predictions. The behavior of the model components are scrutinized and means to embellish the model to account for muscle state and trunk velocity are discussed. These findings have implications for ergonomic designs, manual materials handling, standard setting and clinical evaluation.

INTRODUCTION

Biomechanical modeling has become an increasingly popular tool for estimating loads upon the lumbar spine. Accurate biomechanical modeling techniques would significantly aid in the evaluation of job and workplace design as well as provide a practical clinical evaluation technique. The evaluation of loads upon the lumbar spine is a rather difficult task. Direct invasive measurement techniques are expensive and dangerous, especially when motion is involved. For this reason, several biomechanical analysis techniques (models) have been developed which can predict loads on the spine based upon the Newtonian laws of physics applied to the biological mechanism. The benefits of a model which is capable of evaluating loads upon the spine would be substantial. Such a model would make ergonomic evaluation of a workstation possible quantitatively. A model would also be valuable for clinical analysis as well as accident evaluation.

A major concern in biochemical assessment is the loading of the internal structures of the trunk. This is important because these internal structures counteract external moments imposed upon the trunk during activities such as lifting. However, the close proximity of the internal structures to the spine compared with the location of the object lifted, relative to the spine, often causes excessive force to be generated by the internal structures. These internal forces are often the primary loading components of the spine. If these spine forces are understood for a particular task, then they can be compared to spine tolerance limits, and the risk of workplace factors can be assessed. Thus, a desirable feature of a biomechanical model of the trunk is that it should predict both the synergistic

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action of the various internal forces acting upon the spine as well as the loading due to external forces.

One of the few biomechanical models which accounts for the activity of both the external forces as well as the collective co-activations of the internal structures is the transverse plane analysis technique proposed by Schultz and Andersson (1981). This model predicts the internal activity of 10 trunk muscles and intra-abdominal pressure in response to external forces acting on the trunk. The model has been validated under several static conditions (Schultz et al., 1982a, 1982b, 1983; Andersson et al., 1983; Schultz, 1986). However, most occupational circumstances require trunk motion to occur in order to perform the task. Although many dynamic biomechanical models exist (i.e. Frievalds et al., 1984; McGill and Norman, 1985, 1986; Gracovetsky et al., 1981) few account for the collectives co-activation of the internal forces within the trunk in response to external forces in a clearly predictable fashion. Schultz and Andersson (1981) suggested that trunk motion could be accounted for by using the model in a quasi-static fashion yet this condition has not been tested. Therefore, the objective of this study was to introduce controlled motion as a variable into the model and determine if the quasi-static assumption can be used to predict the activities (in particular the co-activities) of the internal forces.

Schultz and Anderssons' analysis technique assumed that the body could be considered divided into an upper and lower segment by an imaginary transverse cutting plane. This cutting plane was passed through the level of the lumbar spine at which the loads were to be assessed. Next, the Newtonian laws of physics were applied to the upper section of the transverse plane in a two step process. The first step involved the calculation of the net reaction. This step was not necessary in the present situation since the net reaction was measured directly with an isokinetic dynamometer. The second step involved the estimation of the internal forces which counteracted the net reaction. This step was the major concern in this study.

Schultz et al. (1982a, 1982b, 1983, 1986) validated their analysis method based upon lengthy (10–15 s) static exertions and hypothesized that the model could describe dynamic situations by describing the task under quasi-static conditions. A similar analysis method has been employed by Chaffin and Baker (1970) and NIOSH (1981).

The experiment performed in the present study involved isometric and isokinetic motion. Pure isokinetic motion involves no acceleration. The present experiment ensured this was the case by allowing a 22.5 degree range of angle for acceleration purposes prior to the analyzed portion of the data. Hence, all acceleration components in the experiment had terminated by the time the data acceptance portion of the experiment had begun.

The analysis procedure developed by Schultz and Andersson predicted spinal compression as well as the trunk muscle forces and intra-abdominal pressure. Spinal compression was not monitored in the experiment. However, internal force generation was recorded by observing the electromyographic activities of the trunk muscles and the intra-abdominal pressure. These parameters are known to relate to force and pressure, respectively. These recordings can then be compared to the analysis predictions. Furthermore, these predictions could be compared with both the isometric exertions, which have been validated, and isokinetic exertions, which have not been validated.

The model

In order to estimate the internal forces, according to the Schultz and Andersson method, the physical arrangement of the transverse cutting plane through the body must be considered. The ten muscle equivalent vectors which represent the major muscle groups of the lumbar region would be identifiable along this plane and would represent muscle internal forces. Each muscle group is oriented with respect to a central axis aligned in the transverse plane. There are five muscles on the right side of the axis center. These are the latissimus dorsi equivalent, L_r, the right erector spinae equivalent, E_r, the right external oblique muscle equivalent, X_r, the right internal oblique, I_r, and the right rectus abdominus equivalent, R_r. There are also five corresponding muscle equivalents symmetrically oriented on the left side of the axis center. An intra-abdominal pressure force, P, has also been incorporated into the analysis. This equivalent force was assumed to act at an equivalent point within the abdomen. Three spine motion-segment resistances were also included in the analysis. These resistances consisted of a compression force C, a right-lateral shear force S_r, and an anterior shear force S_a. These three motion-segment resistances were assumed to act at the coordinate system origin. A detailed description of these variables may be found in Schultz et al. (1981).

The net force, which must be counterbalanced by the internal forces, consisted of three net force components and three net moment components. These force components were also aligned with the coordinate system described. These forces were assumed generated by the portion of the body above the imaginary transverse cutting plane. The reaction to these forces, needed to keep the body in equilibrium, must then be supplied by the action of the internal forces of the body below the imaginary transverse cutting plane. If the Newtonian laws of physics are observed, the requirement that the 14 internal forces provide the six components of the net reaction that are needed for equilibrium may be derived. The equations describing this situation can be found in Schultz and Andersson (1981).

Schultz and Andersson suggested that the intra-abdominal pressure may be determined experimentally. That left 13 unknown internal forces and only six available equations for their solution. This situation led to a statically indeterminate problem. Schultz and Andersson suggested two methods to solve this problem. First, they suggested that if enough assumptions about the internal forces were made this would render a statically determinate solution to the problem. They suggested several assumptions such as: (1) no significant intra-abdominal pressure was present, (2) any activity of the latissimus dorsi could be included in the erector equivalent muscles, and (3) antagonistic muscle contractions were minimal. These approaches to solving the problem of indeterminacy appeared inappropriate since the experimental portion of this study revealed significant intra-abdominal pressures and definite differences in the use of the latissimus dorsi and erector spinae muscles as well as the significant co-activation of the antagonistic muscles.

The second method Schultz and Andersson suggested to handle the problem of indeterminacy, was to apply optimization techniques to the problem. The concept of optimization was known to relate well to physiologic functions of the human body. For example, Hatze (1977) used optimization to model the fine control motion of a gait step with good results. Ayoub (1971) found optimization techniques useful in the modeling of the upper extremity, as did Seireg and Arvikar (1975) and Davy and Audu (1987) in their attempts to model the forces produced in the lower extremity. Gracovetsky et al. (1981) also used optimization to predict the influence of spine supporting structures such as ligaments, tendons and muscles. Hence, optimization techniques appeared to provide a justified means to deal with this problem of indeterminacy without having to make unjustified assumptions.

Linear programming model of internal forces

Linear programming provides a direct and simple method of modeling the contribution of the internal forces in response to a net force of the upper body (which is the situation during a lift). This method was chosen to model the activity of subjects attempting to perform the experimental task of this study.

A major concern of linear programming procedures was the selection of an appropriate objective function or quantity to be optimized. In this study, the compression component, C, of spine motion-segment resistances was selected as the quantity to be minimized. Schultz et al. (1982a) investigated several objective functions and found that compression worked well under static conditions. Disc compression has also been identified as a factor which causes end plate fracture (Garg and Herrin, 1979). This component appeared in the equilibrium equation which described the force in the z-axis. This equation was used as the objective function for the linear programming model.

The constraints of the model were divided into two types of constraints. The first type of constraint forced the model to obey the equilibrium equations derived by Schultz and Andersson. In the present study the experimental task required the subjects to perform a sagittally symmetric exertion. Therefore, this task was modeled so the moment about the x-axis, M_x, was equated to the instantaneous torque the subject produced under each experimental condition. Since the experimental task was a sagittally symmetric one, the moments about the y and z axes, M_y and M_z, should be zero. Hence, the M_y and M_z values of the equilibrium equations were equated to zero in the development of the linear programming model.

The second type of constraint placed upper and lower bounds upon the internal forces of the body. The lower bound for each force was a nonnegativity constraint. These nonnegativity constraints required that the force of each muscle and intra-abdominal pressure be at least zero. The upper bounds of the internal forces required that the force contribution of each muscle not exceed a reasonable level, and that the pressure produced in the abdominal cavity not exceed the highest experimental value observed for each subject. The upper bounds for the muscles were determined as a function of the cross-sectional area of each muscle. Based upon previous studies (Ledley et al., 1977) the cross-sectional areas of each muscle at the L_4 level was computed from computer tomography (CT scan) pictures. These cross-sectional areas were then multiplied by the maximum expected contraction intensity of human muscle (100 N/cm^2, as suggested by Schultz and Andersson (1981)).

Together, the objective function, the moment controlling equations and the upper and lower boundary constraints for the internal forces, constitute the linear programming model of the experimental task. The formulation of this model is shown below:

$$\text{MINIMIZE } C = -P + E_L + E_R + R_L + R_R + I_L(0.707) + I_R(0.707) + L_L(0.707) + L_R(0.707) + X_L(0.707) + X_R(0.707)$$

SUBJECT TO:

$$(1)\ 4.4E_L + 4.4E_R - 10.8R_R - 10.8R_L + 4.8(P) + (5.6L_L + 5.6L_R)0.707 - 3.8(I_L + I_R)0.707 + (X_L + X_R)0.707 = TORQUE$$

$$(2)\ 5.4E_R - 5.4E_L + 3.6R_R - 3.6R_L + 4.45L_R - 4.45L_L + 13.5(I_R - I_L)0.707 + (X_R + X_L)0.707 = 0$$

$$(3)\ (5.6L_R - 5.6L_L)0.707 + 13.5[(I_R - I_L)0.707 - (X_R - X_L)0.707] = 0$$

(4) $P \geqslant 0$	(15) $P \leqslant$ Exp. value
(5) $E_L \geqslant 0$	(16) $E_L \leqslant 1800$
(6) $E_R \geqslant 0$	(17) $E_R \leqslant 1800$
(7) $R_L \geqslant 0$	(18) $R_L \leqslant 1200$
(8) $R_R \geqslant 0$	(19) $R_R \leqslant 1200$
(9) $L_L \geqslant 0$	(20) $L_L \leqslant 240$
(10) $L_R \geqslant 0$	(21) $L_R \leqslant 240$
(11) $I_L \geqslant 0$	(22) $I_L \leqslant 1000$
(12) $L_R \geqslant 0$	(23) $I_R \leqslant 1000$
(13) $X_L \geqslant 0$	(24) $X_L \leqslant 800$
(14) $X_R \geqslant 0$	(25) $X_R \leqslant 800$

The experiment

The experiment involved sagittally symmetric isometric and isokinetic "back-lift" maximum exertions. Twenty subjects were monitored while they attempted to exert torque about the L5/S1 junction of the back as is done during lifting. The instantaneous activity of the 10 trunk muscles, intra-abdominal pressure (IAP) and torque produced by the back were evaluated as subjects passed through three trunk angle (0°, 22.5° and 45° forward angle) conditions under one of the velocity conditions (0% (isometric), 33%, 67% or 100% (isokinetic) of maximum velocity). The activities of the ten trunk muscles were monitored via electromyography (EMG) using fine wire electrodes implanted in the muscles of interest. The integrated EMG activity was used to represent muscle force. Muscle force was normalized for all muscles by dividing each EMG value by the maximum EMG value observed during maximum trunk exertions. IAP was measured using a catheter-transducer inserted into the stomach via the nasal passage. Details of the experiment are reported elsewhere (Marras et al., 1984).

RESULTS

Model Performance

A model run was performed for each subject under each experimental condition. The only unmeasurable quantity in the experiment which was predicted by the model was spinal compression. Summary results of these predictions are shown in Fig. 1. As can be seen, spinal compression increased as a function of increasing angle and decreased as a function of increasing velocity.

The performance of linear programming models is such that the model moved from boundary point to boundary point and tested the objective function at each point until the optimal value of the objective function was found. In the present situation, the internal forces with the greatest mechanical advantages were the variables which appeared in the basis of the solution. For this reason, the latissimus dorsi muscles were always entered into the solution and they were almost always employed at the maximum possible level. The predicted activities of the internal forces were converted to the percentage of the maximum activity seen in that internal force for each subject, so they could be compared with experimental performance.

The only other muscles which were chosen by the model as a solution to the problem were the erector spinae right and left muscles. These muscles were the set of muscles with the second greatest mechanical advantage to enter the solution. However, unlike the latissimus dorsi muscles, these muscles did not consistently achieve maximum utilization.

Finally, the only other internal force predicted by the model was the intra-abdominal pressure activity. This variable's mechanical advantage was the least of those in the basis.

Model–Experiment comparison

The model predictions were compared with the experimental data where applicable. The internal forces which were available from both model prediction and experimental measurement were the activities of the latissimus dorsi muscles, erector spinae muscles and intra-abdominal pressure. All of these quantities were transformed into percentage values for comparison purposes.

A summary of model predictions and experimental measurements appears in Table 1. The model predicted greater usage of both the latissimus dorsi muscles than were experimentally measured. The model predicted an average activity of about 93 percent of the maximum whereas the average activity of these muscles was measured at about 59 and 48 percent of the maximum for the right and left side, respectively. Standard deviations for these values were observed to be relatively similar. The relative difference between the model predictions and experimental data of the latissimus dorsi muscles could also be contrasted

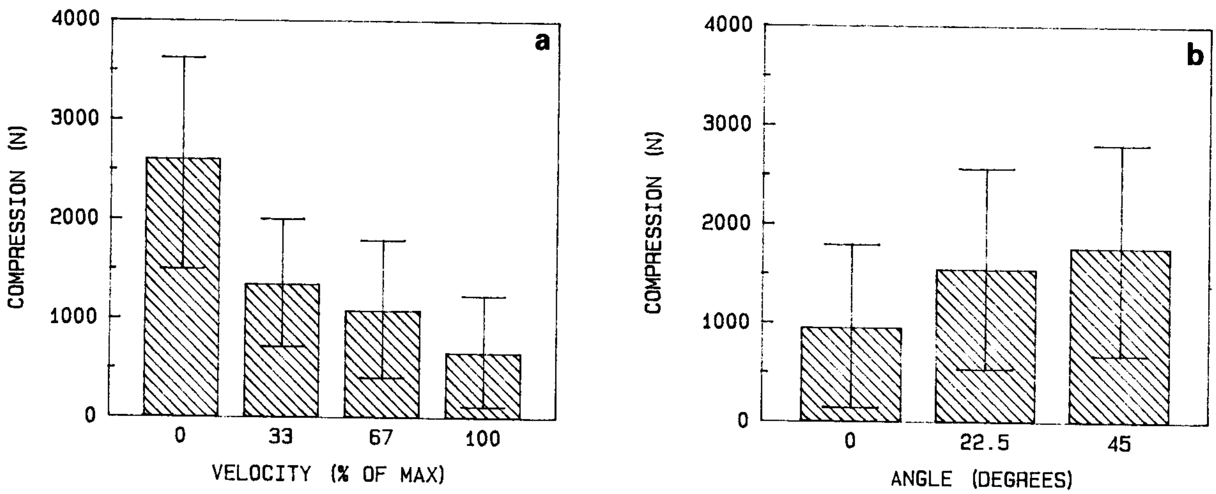

Fig. 1. Model predictions (means and standard deviations) of compression as a function of (a) velocity and (b) angle.

TABLE 1

Experiment–Model comparison

		Experimental data		Model data
LATR	$\bar{x}$	58.65		92.87
	s	22.53		20.69
Correlation coefficient (r)			0.170	
LATL	$\bar{x}$	47.85		92.87
	s	23.38		20.64
Correlation coefficient (r)			0.153	
ERSR	$\bar{x}$	55.64		39.14
	s	25.64		33.39
Correlation coefficient (r)			0.374	
ERSL	$\bar{x}$	56.47		39.13
	s	27.84		33.39
Correlation coefficient (r)			0.366	
IAP	$\bar{x}$	53.21		66.28
	s	23.75		34.50
Correlation coefficient (r)			0.564	

by the comparisons shown in Fig. 2. These comparisons indicated a substantial model overprediction of activity for the latissimus dorsi muscles in response to both the velocity and angle factors. Furthermore, the relative response characteristics of these muscles was not similar. Experimental data indicated linear and cubic response to the velocity factor throughout the experimental range, whereas, no such cubic components were seen throughout the model predictions. The association between responses to the angle factor for this muscle group was only slightly better. The model still overpredicted the muscle activity but did indicate a quadratic component in the LATL muscle, as was observed in the data. However, the nature (sign) of the quadratic component was exactly opposite to that observed for the experimental data. Correlation coefficients were calculated be-

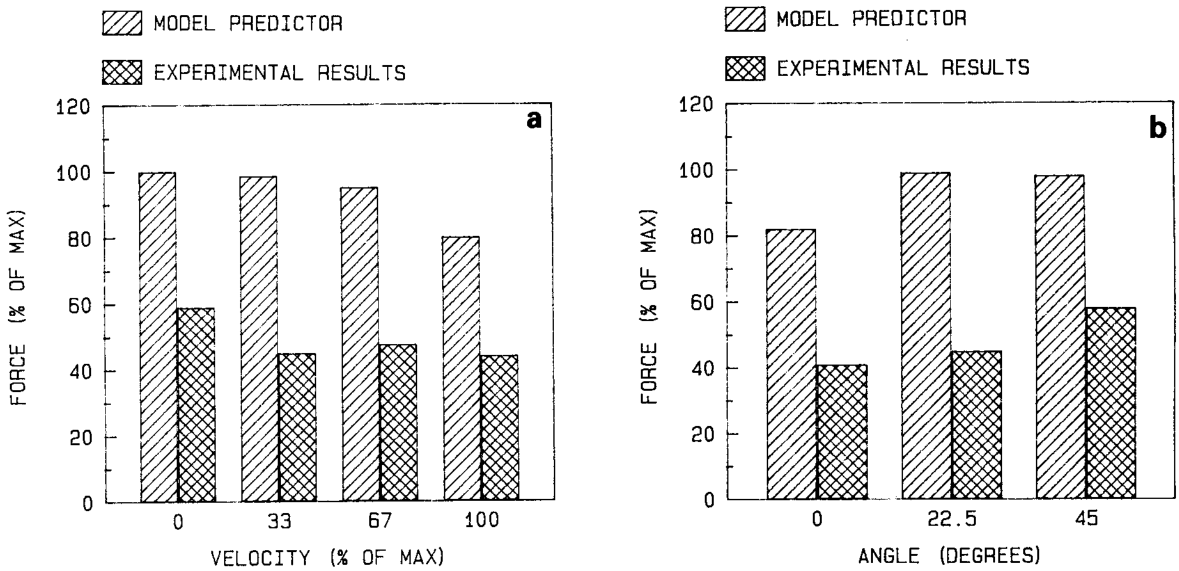

Fig. 2. Model predictions vs. experimental recordings of latissimus dorsi activity as a function of (a) velocity and (b) angle.

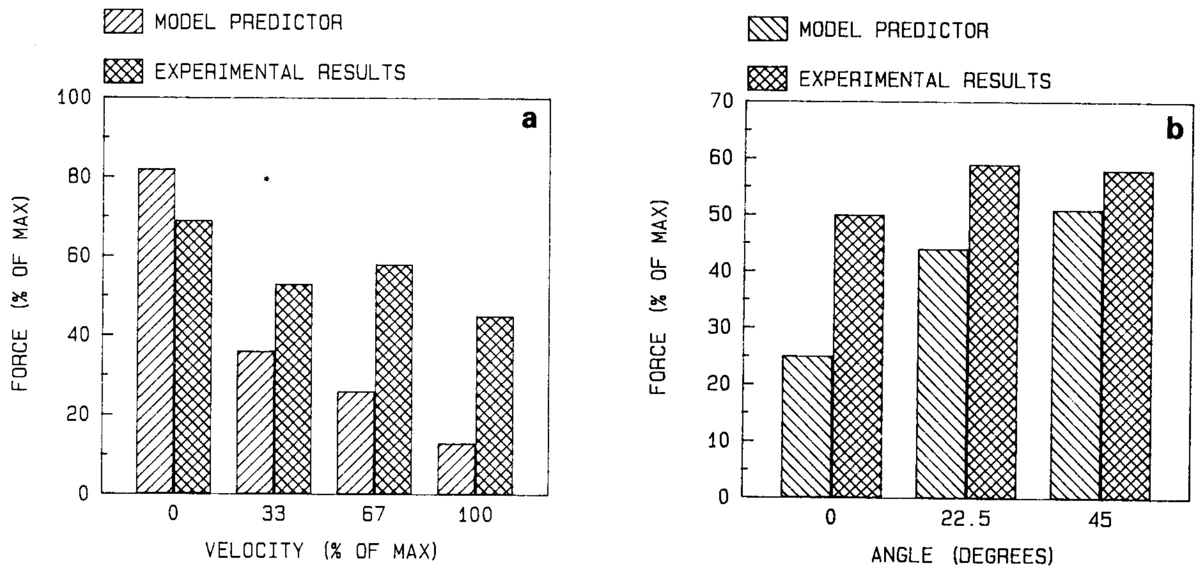

Fig. 3. Model predictions vs. experimental recordings of erector spinae activity as a function of (a) velocity and (b) angle.

tween model prediction and measurement findings. Neither the LATR nor LATL exhibited significant linear correlations ($r_{LATR} = 0.170$, $r_{LATL} = 0.153$, d.f. = 240, $p > 0.005$).

The relationship between the model prediction and experimental observation for the erector spinae muscle was much greater. The average activity of these muscles was under-predicted by about 16 percent via model prediction as compared to the actual experimental data. However, standard deviations indicated the model predicted more variability than actually occurred by experimental measurement. The relative difference between model prediction and the experimental differences can also be visualized for this muscle group by the comparisons of Fig. 3. In terms of the response to velocity conditions, not only did the model predict activity much closer to the

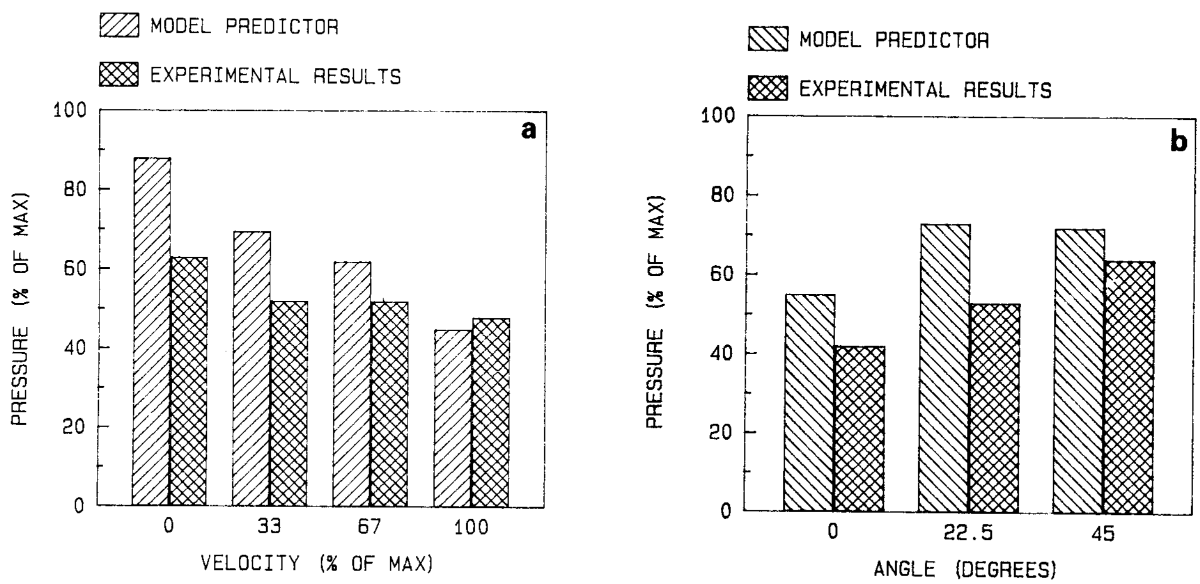

Fig. 4. Model predictions vs. experimental recording of intra-abdominal activity as a function of (a) velocity and (b) angle.

recorded activity, but the relative trend component shape response was much closer. Here the model overpredicted the V0 condition and underpredicted the V33, V67 and V100 conditions. Linear and cubic components were found for the experimental data, whereas only linear components were seen for the model. The response between angle conditions also showed similar differences. Quadratic components were evident in the data which were not seen in model predictions. A significant linear correlation was observed between model predictions and experimental observation for both the ERSR and ERSL variables ($r_{\mathrm{ERSR}} = 0.374$, $r_{\mathrm{ERSL}} = 0.366$, d.f. = 240, $p < 0.005$).

Finally, the activity of the intra-abdominal pressure was compared (Fig. 4). Here again, the model overpredicted this activity by about 16 percent on the average. Also, the greatest difference in variability was seen here via the standard deviation. The difference in IAP prediction and measurement can be seen via the comparisons. The experimental data showed that the IAP response to velocity was much less reactive to increases in velocity, than was predicted by the model. Actual data also indicated a much more linear response to increasing angle than did the model. Even with these differences, the highest significant linear association between predicted and observed measures was noted for this variable. The correlation coefficient was the greatest of all variables for intra-abdominal pressure ($r_{\mathrm{IAP}} = 0.564$, d.f. = 240, $p < 0.005$).

DISCUSSION

Even though Schultz and Andersson have validated their model, several qualifying aspects of the validation process should be kept in mind when assessing the merits of their biomechanical analysis technique. First, the validation process used by Schultz and Andersson employed surface EMG electrodes for all electromyographic recordings. This technique made it impossible to isolate the electromyographic activity of individual muscles. The sum of all the activity under the electrode was recorded. Hence, muscles which overlap each other were not distinguishable.

A potential difference in the manner in which the model was formulated in this study involved the determination of the upper limit constraints of the trunk muscles in the model. These upper limits were a function of the cross-sectional area of the muscles at the selected level. The cross-sectional areas for this study as well as previous validation studies of this model were derived from a book of computer tomography pictures and were not necessarily representative of the subject population tested in the experimental portion of this study. The only way to properly validate the model, with respect to these limits, would be to take a CT scan of each subject and tailor the model to these individual muscle variations.

An inherent shortcoming of the model when used to predict sagittally symmetric lifting behavior, is that there remain only three functionally based equations in the model constraints under these conditions. Due to the structure of linear programming there is only a capability for at most three variables to take on non limit values. Hence, the model predicts zero for all non-back trunk muscles. More functional relation equations are needed for the model to behave any other way.

The effects of this shortcoming are realized in the inability of the model to predict the activity of the internal oblique, external oblique and rectus abdominus muscles. The experimental results indicate that these muscles were active under all experimental conditions. However, the model predicted that these muscles would be inactive. The forces generated by these muscles are important since they represent a co-activation of trunk supporting structures. Therefore, when structures provide force simultaneously we would expect an "impulse" load to the spine. Accordingly, the model would erroneously estimate spine compression under the experimental conditions. Co-activation of the muscles influence instant loading.

Generally, the model was capable of predicting the activity of the erector spinae right and left muscles and the intra-abdominal pressure activity within a mean range of 16% of the experimental value. However, the predictive capability in terms of the latissimus dorsi muscle activity was less than impressive and on the order of a 30 to 35% mean overprediction. The experimental portion of this study showed that the latissimus dorsi involvement in the production of torque changed significantly as a function of the angle of the back. This property was probably due to the fact that the contractile potential of muscle was a function of the relative state of the muscle. A muscle which is in a shortened state does not have the power producing capability that the same muscle would have in a somewhat lengthened state. This is a well documented fact for many muscles of the extremities. For example, the elbow is known to have the greatest power potential at a 90 degree angle. Hence, the muscle force should be a function of the relative length of the muscle besides just the cross-sectional area of the muscle. This adjustment could be performed with equations which describe the power producing potential of the muscle. This would also make the model more sensitive to angle and provide the potential for more non-limit solutions.

Another inherent shortcoming of a linear programming model applied to a human system, is that it is based upon linear relationship between model parameters. The regression analysis of the experimental data revealed significant curvilinear components of some of the internal structures of the body. These nonlinearities may also be present in the functional relationship of the internal mechanisms. The model analysis technique does not account for these components and their effect on spine impulse loading in its present form. Equations which describe how the trunk muscles respond to motion are required. The basic structure of the force and moment equations used in the model appear sound. However, means are needed to describe the varying state of the muscles and their subsequent influence on spine loading during trunk motion. This would also help add equations to the model which could lead to statically determinate solutions.

CONCLUSION

This study has compared a trunk loading biomechanical prediction technique with the results of an experimental study of trunk structure response to motion. Even though previous studies have shown that the prediction model works well for static exertions, significant differences were identified for trunk motion conditions. This study has identified these differences and has suggested how the model may be changed to account for trunk motion. Such changes would increase the ergonomic usefulness of the model for workplace purposes.

REFERENCES

Andersson, G.B., Ortengren, R., Nachemson, A.L. and Schultz, A.B., 1983. Biomechanical analysis of loads on the lumbar spine in sitting and standing postures. In: H. Matsui and K. Kobayashi (Eds.), Int. Series on Biomechanics. Human Kinetics Publishers, Champaign, IL, pp. 543–552.

Ayoub, M.A., 1971. A biomechanical model for the upper extremity using optimization technique. Ph.D. Dissertation, Texas Tech. Univ.

Chaffin, D.B. and Baker, W.H., 1970. A biomechanical model for analysis of symmetric sagittal plane lifting. AIIE Trans., II (1): 16–27.

Davy, D.T. and Audu, M.L., 1987. A dynamic optimization technique for predicting muscle forces in the swing phase of gait. J. Biomech., 20(2): 187–201.

Freivalds, A., Chaffin, D., Garg, A. and Lee, K., 1984. A dynamic biomechanical evaluation of lifting maximum acceptable loads. J. Biomech., 17(4): 251–262.

Garg, A. and Herrin, G.D., 1979. Stoop or squat: A biomechanical and metabolic evaluation. Amer. Inst. Ind. Eng., 11: 293–302.

Gracovetsky, S., Farfan, H.F. and Lamy, C., 1981. The mechanism of the lumbar spine. Spine, 6(3): 249–262.

Hatze, H., 1977. A myocybernetic control model of skeletal muscle. Biol. Cybernetics, 25: 103–119.

Ledley, R.S., Huang, H.K. and J.C., Mazziotta, 1977. Cross-sectional Anatomy—An Atlas of Computerized Tomography. Williams and Wilkins, Baltimore, MD.

Marras, W.S., King, A.I. and Joynt, R.L., 1984. Measurements of loads on the lumbar spine under isometric and isokinetic conditions. Spine, 9(2): 176–188.

McGill, S.M. and Norman, R.W., 1985. Dynamically and statistically determined low back moments during lifting. J. Biomech., 18(12): 877–885.

McGill, S.M. and Norman, R.W., 1986. Partitioning of the L4-L5 dynamic moment into disk, ligamentous and muscular components during lifting. Spine, 11(7): 666–678.

NIOSH Technical Report, 1981. Work practices guide for manual lifting. DHHS (NIOSH) Publication No. 81–122, March.

Schultz, A.B., 1986. Loads on the lumbar spine. Mech. Eng., 108(1): 36–41.

Schultz, A.B. and Andersson, G.B., 1981. Analysis of the loads on the lumbar spine. Spine 6(1): 76–82.

Schultz, A.B., Andersson, G.B., Ortengren, R., Nachemson, A., Haderspeck, K., 1982a. Loads on the lumbar spine: Validation of a biomechanical analysis by measurements of intradiseal pressures and myoelectric signals. J. Bone Joint Surg., 64-A(5): 713–720.

Schultz, A.B., Andersson, G.B., Haderspeck, K., Ortengren, R., Nordin, M. and Bjork, R., 1982b. Analysis and measurement of lumbar trunk loads in tasks involving bends and twists. J. Biomech., 15(9): 669–675.

Schultz, A.B., Haderspeck, K., Warwick, D. and Portillo, D., 1983. Use of lumbar trunk muscles in isometric performance of mechanically complex standing tasks. J. of Orthop. Res., 1: 77–91.

Seireg, A. and Arvikar, R.J., 1975. The prediction of muscular load and joint forces in the lower extremities during walking. J. Biomech., 8: 89–102.