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A Three-Dimensional Motion Model of Loads on the Lumbar Spine: I. Model Structure

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Traditionally most biomechanical models that are used to estimate the loading experienced by the spine during work focus on static, two-dimensional representations of the work. However, most work tasks impose loads on the lumbar spine under dynamic, three-dimensional conditions. The objective of this study was to describe the structure and logic of a model that is capable of producing estimates of spine loading under three-dimensional motion conditions. This model is intended for use primarily under laboratory conditions. The model was designed initially for workplace simulation in which the trunk is moving under symmetric and asymmetric constant velocity lifting conditions. Future embellishments may enable the model to be used under free dynamic conditions. The model predicts lumbar spine compression, shear, and torsional forces as well as trunk torque production continuously throughout the exertion. This information may be compared with spine tolerance limits so that the risk of causing a vertebral end-plate microfracture by workplace requirements could be determined.

INTRODUCTION

Models that represent loading of the spine in occupational circumstances are essential for proper assessment of the work environment. Typically model estimates of spine loading during the performance of a task are compared with tolerance limits of the vertebral bodies. In this manner one could determine how workplace factors and methods influence both the acute and cumulative loading of the spine throughout the workday.

One of the real challenges in such efforts involves the predictions of loads on the lumbar spine under realistic occupational conditions. The literature (e.g., Kroemer, Snook, Meadows, and Deutsch, 1988) is beginning to reflect the appreciation within the field for the differences between internal and external loading of the body. The occupational biomechanics literature assumes that external as well as internal stresses (forces) are acting on the body during a manual materials handling (MMH) task. The external forces produce moments around the spine attributable to the mass of the object lifted and the weights of the body segments and their distances from the spine. The internal forces are those supplied by the muscle forces, pressures, and passive components within the body in order to provide a countermoment to the external

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moments. However, because of their location in the body, the moment arms (lever arms) of these forces are much shorter than those of the external forces. Thus these muscles are at a severe mechanical disadvantage and must produce much larger forces in order to counterbalance the external moments.

Most knowledge concerning the reaction of the trunk internal spine loading structures (muscles, intraabdominal pressure, ligaments, and other connective tissue) has been based on static exertions of the trunk. Chaffin and Baker (1970) were the first to create a model of loading on the lumbar spine under occupational conditions. Their model, which was two-dimensional and static, consisted of several links representing the major articulations of the body. They computed the torques imposed on each joint and the subsequent compression on the spine. The effects of internal trunk loading were not considered other than from trunk extensor muscles represented by a single equivalent trunk muscle. Schultz and Andersson (1981) later created a lumbar spine model that could be used for workplace assessment. Their model considered the effects of external moments (imposed around the spine) on the activity of 10 trunk muscles and on intraabdominal pressure. This model computed compression and shear forces on the spine primarily under static conditions, but it was not designed to calculate additional spine loading attributable to coactivation of the trunk muscles.

Examinations of most typical industrial workplaces indicate that jobs in those environments often require dynamic trunk motion. Several dynamic models and assessments appear in the literature (Freivalds, Chaffin, Garg, and Lee, 1984; Leskinen, Stalhammar, Kuorinka, and Troup, 1983; Mital and Kromodihardjo, 1986; Smith, Smith, and McLaughlin, 1982; Wood and Hayes, 1974). Many of these models have estimated that loading of the lumbar spine under dynamic conditions is between 22.5% and 60.0% greater than loading under static conditions. Most of these models were based on kinetic or kinematic body motion information. The researchers computed the spine loads attributable to inertial forces generated by the body segments during motion as well as spine loading from static gravitational forces. One problem with this approach is that many of these models did not include the action or coactivation of the trunk musculature. We assume that internal force magnitudes become even greater and coactivation of agonist-antagonist pairs of muscles would become more frequent with trunk motion because the inertial moment of the trunk must also be overcome. Both of these events would be expected to increase spine loading significantly. Another problem with this approach is that most of these models are complex and lack the simplicity that would make them valuable for routine workplace evaluation.

McGill and Norman (1985) developed a model of spine loading during sagitally symmetric trunk motion using electromyographic (EMG) data as input, thereby accounting for muscle coactivity. This model incorporated previously published relationships of internal structure responses to motion conditions in order to assess spine loading while tracking body segment motion. However, limited empirical evidence of the model's validity was provided. It seems that many of the models that are available for assessment of spine loadings are still in an evolutionary stage and therefore have not been useful in the assessment of loading on the lumbar spine under realistic working conditions.

These studies have shown that the effects of dynamic motion on the loading of the spine are not fully understood. We believe a major void exists in the field in light of the paucity of models that attempt to evaluate dynamic spine loading caused by the coactivity of the trunk musculature during motion. Several studies have reported significant coactivity during trunk motion (Marras and Mirka, 1990, in press; Marras and Reilly, 1988). Most predictive models that attempt to account for the effects of multiple trunk muscles—for example, the Schultz and Andersson (1981) model—yield indeterminate solutions, which require that some muscles be inactive, though this is contrary to empirical findings. Therefore, assumptions regarding a lack of muscle coactivity or optimization modeling techniques are often used to facilitate a solution. Research has shown that the assumption of a lack of coactivity is not justified under dynamic conditions (Marras and Mirka, 1990, in press; Marras and Reilly, 1988). Optimization models usually cannot account for coactivity because the quantity of model solutions is limited by the number of functional constraints. These functional constraints are usually derived from six trunk force and moment equations (Schultz and Andersson, 1981). Hence it is not possible to determine more than six nonboundary solutions, or six of the muscle forces. The optimization models that have managed to include coactivity have forced it to occur by imposing artificial model boundary constraints (see Marras, 1988). Consequently, these models cannot predict the same level of coactivity that is observed experimentally.

It is clear from this review that we are only beginning to understand the influence of dynamic trunk motion on the activity of the trunk musculature. Marras and Mirka (1990, in press) have studied the reactions of the trunk musculature to controlled trunk motions that are similar to those seen in the workplace. They described the behavior of each of 10 important trunk muscles in response to changes in trunk velocity and trunk acceleration. Significant coactivation among trunk muscles was observed in these studies. However, coactivity increased differently depending on whether the trunk increased its velocity or acceleration. This coactivity surely influences the loading of the lumbar spine, yet few models consider the effects of this coactivity on the loading of the lumbar spine. Predictive models have been unable to account for this coactivity and therefore may underestimate the loading on the spine.

Another difficulty with predictive models is that they are seldom capable of accounting for individual variability. Most predictive models use objective functions that optimize spine loading or account for body kinetics in the most efficient manner. These models predict the same loadings for all subjects under a particular condition. However, the trunk musculature behaves in a highly individualized manner when creating trunk motion, and this variability greatly influences spine loading.

In order to overcome these problems we have developed a descriptive simulation model of trunk loading under dynamic trunk motion conditions. This model is intended to be a tool that allows one to evaluate the collective influence of the trunk musculature on spine loading as the trunk moves under controlled laboratory conditions. Laboratory conditions are necessary because information about trunk muscle activity must be recorded via electromyography, and trunk torque and posture must also be monitored. Marras and Reilly (1988) have described such an approach and found that it was very sensitive to trunk motion changes. They were able to describe the sequence of internal muscle activations (Marras and Reilly, 1988) as well as estimate relative spine loading (Reilly and Marras, 1989) using a simulation model. In the present paper we have expanded on this approach in two ways. First, we assessed absolute (as opposed to relative) spine loading in order to predict actual loads experienced by the spine. Second, we accounted for trunk asymmetry during controlled dynamic motion of the trunk. The goal throughout the model development process has been to understand the degree to which the mechanical aspects of muscle contraction influence trunk motion and loading. This strategy has permitted us to increase the complexity of the model only when significant improvements in accuracy could be realized.

This model can be used to facilitate exploratory research of trunk loading during the production of trunk velocity and torque. Such a model would make it feasible to vary the amount of trunk velocity and trunk torque independently in order to investigate the subsequent changes in spine loading systematically. Through systematic study we could improve our understanding of how an individual's trunk copes with each of these changes, without the confounding that occurs when both factors are permitted to vary simultaneously.

In summary, the objective of this paper is to describe a model that utilizes muscle activity information in order to completely describe the collective influence (in terms of amplitude and temporality) of the trunk musculature on the loading of the spine during trunk motion.

MODEL DEVELOPMENT

Goals

Goals for the development of this model included the ability to examine trunk muscle activity patterns and to quantify three-dimensional stresses on the spine throughout the duration of an isokinetic trunk lifting exertion. An additional goal was to develop a model that would be complex enough to give reasonable solutions while remaining simple enough to be used by individuals other than those who developed it.

Model Overview

When attempting to model dynamic exertions and the resultant internal spinal loadings, it is necessary to estimate muscle force. In order to estimate the amount of force generated by a given muscle at any point in time during the exertion, it is essential to consider the physiological aspects of the muscle's response to externally imposed torques applied over time. Figure 1 displays the basic features of the logic within the model for incorporating muscle physiology and trunk motion components. The key model inputs are categorized as subject characteristics, EMG signal characteristics, and trunk kinematic and kinetic elements. Subject characteristics refer to subject anthropometry, which is utilized in estimating muscle cross-sectional areas and trunk weight. Two aspects of the integrated EMG signal are employed: the amplitude and, because of the dynamic nature of the exertion, the temporal information in the signal. In this way the effects of peak muscle exertions and muscle coactivity can be captured and their contributions to spinal loading can be determined. Another necessary ingredient for greater understanding of dynamic exertions is information pertaining to trunk kinematics and kinetics. This includes trunk flexion angle, angular velocity, and the output torque that the trunk is imparting to offset any external load. These elements affect one or more aspects of muscle physiology, and including their effects leads to an improved understanding of muscle response and internal spinal loading throughout the dynamic exertion.

The basic movement to be modeled at this phase of model development was an isokinetic trunk extension performed at moderate speeds (10, 20, or 30 deg/s). Kim and Marras (1987) found that these trunk velocities were typical of velocities attained in MMH tasks. The model currently operates by using as input EMG data collected during isokinetic trunk extensions performed in the laboratory. These data, after being modulated and synthesized, are used to calculate stresses on the

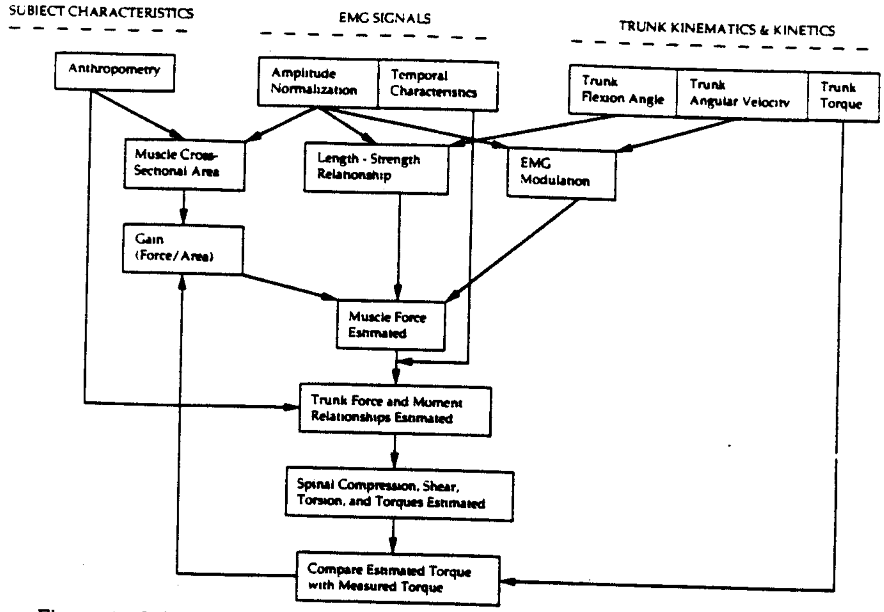


Figure 1. *Schematic of model depicting model inputs and their use in the model.*

spine of the subject throughout the exertion. A goodness check is performed which compares calculated torque around the lateral axis with the torque values measured during the actual exertion.

For an individual subject and specific trial conditions, the model calculates spinal loading attributable to the activity of 10 trunk muscles and the weight of the body above the L5 level. These values are computed throughout the extension and are based on the EMG activity of the 10 trunk muscles during the exertion, force and moment equations, and several assumptions regarding muscle characteristics and EMG activity.

Specific Model Structure

Flow diagram. Throughout this discussion it may be useful to refer to the conceptual flow diagram in Figure 2, which illustrates the inner workings of the model.

Model inputs. The model uses data from isokinetic trunk extension trials performed through the employment of an isokinetic dynamometer, certain anthropometric data, and maximum EMGs for use in normalizing the EMGs for the particular trial of interest. The particular task mimics the motion of the back during an MMH that would involve only the back—that is, lifting is not aided through use of the legs.

Data required by the model include trunk torque, trunk flexion angle, and trunk angular velocity data collected throughout the exertion. The model also requires EMG data from five (left/right) pairs of trunk muscles: the latissimus dorsi, erector spinae, rectus abdominus, and internal and external obliques. All of this information is collected under controlled conditions. It is important to control strictly the point of rotation about which the trunk rotates, the constant angular speed of the trunk, and any trunk twisting (trunk asymmetry) that is intentionally introduced.

Maximum EMG levels pertaining to static maximum exertion trials performed at three angles of trunk flexion are used in the EMG

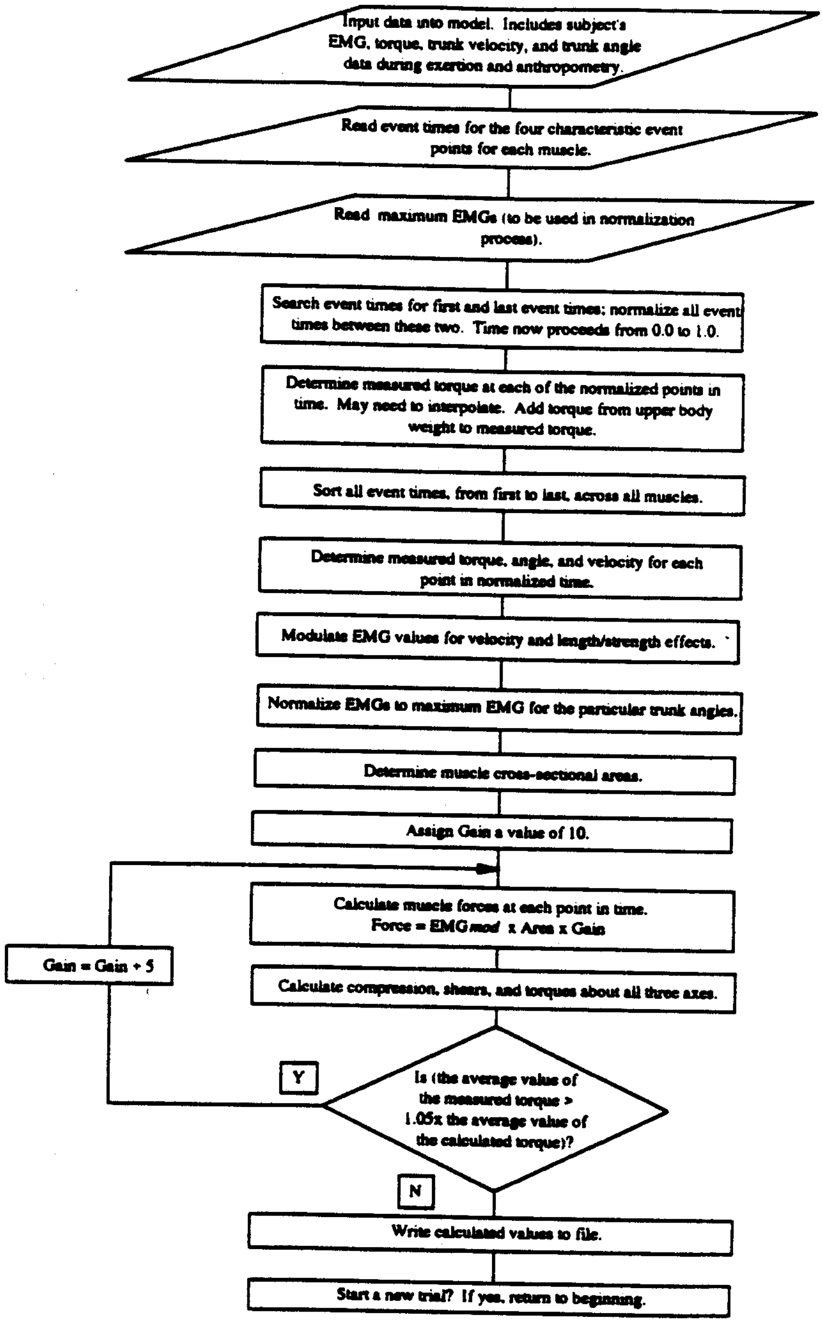


Figure 2. *Model flow diagram.*

normalization process. The maximum value is equated with a maximum level of force generation capacity. Actual trial EMG values are divided by these maxima in order to determine the activity level at which the muscle is operating. As an example, if the maximum capacity is considered to be 300 N and the normalized EMG was 0.35, then the force from the muscle, as indicated by the EMG, would be 105 N (0.35 × 300 N).

Subject anthropometry, which includes height and weight measures as well as torso depth and breadth at L5, is used in the model to determine the moment attributable to the weight of the upper body and in calculating muscle cross-sectional areas and muscle moment arms.

Force from intraabdominal pressure (IAP) was not included because of the minimal effect it probably has in relation to the forces from the muscles under these lower-level external loading conditions. Marras and Mirka (in press, 1990) measured IAP and concluded that there is minimal change in IAP under these conditions.

The other important input to the model is a file (for each exertion trial) containing the characteristic event times for each muscle. For a given trial each muscle's EMG signal is reduced, in a sense, to four event points that can be said to characterize the signal profile. Points 1 and 4 are the points in time that signal the start and end of the activity of that muscle. Points 2 and 3 are any points in between which help to define the signal profile. Refer to Figure 3 for examples of how these points are chosen.

Processing of inputs. Once the aforementioned information is made available to the model, the model then begins to operate on the data. The event times are ordered from first to last across all the muscles, and the time scale is converted from absolute time to relative time, where time 0.0 is the occurrence of the first event (the first muscle activating) and time 1.0 marks the final event (the last muscle returning to resting level).

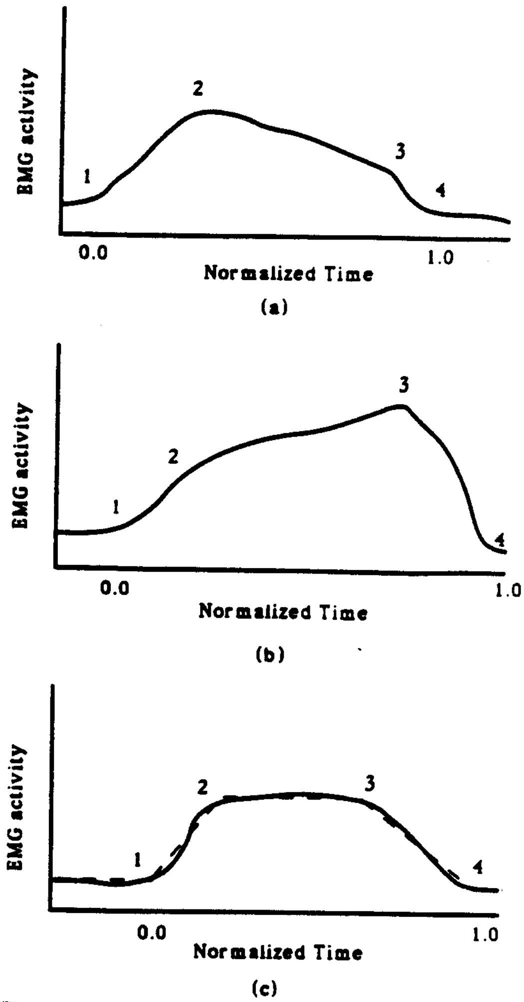


Figure 3. *Figures a, b, and c depict the manner in which the four characteristic points are selected for typical EMG records. Figure c includes the linear approximation which the model assumes.*

Implicit in the notion of reducing the EMG signals to four event points is the idea that the values of the signal in between those points can be calculated using a straight-line approximation (dashed line in Figure 3c).

Based on this assumption, and given the velocity, angle, and torque at each point in time that is marked by an event (related to any muscle), the EMG values for each muscle at each point in time can be estimated. A total of 40 event times per trial are possible, given the four event times for each of the 10 muscles. There may be fewer than 40 distinct event times if some events occur simultaneously or if one or more of the muscles are not recruited during the exertion.

Once an EMG value is established for each muscle at every event time, those values are then modulated by velocity and length-strength effects.

Determining force from the EMG signal. Although each operation is actually performed separately, Equation 1 sums up all of the modifications made to the EMG signal in order to derive a tension level for each muscle at each event point in time.

$$Force = gain \times \frac{EMG}{EMG_{max}} \times V\,ratio \times L\text{-}S\,factor \times area, \qquad (1)$$

where *force* = muscle tension associated with EMG, *gain* = factor that includes maximum muscle force per unit of area, *EMG* = recorded or interpolated value at a particular event time, EMG_{max} = maximum EMG value for particular muscle at particular angle of orientation, *V ratio* = velocity modulation factor, *L-S factor* = length-strength modulation factor, and *area* = muscle cross-sectional area.

The workings of the gain factor, V ratio, and L-S factor, as well as the EMG normalization procedure, are described in detail later in the paper. Only the EMG signals from the latissimus dorsi, erector spinae, and internal oblique muscles are currently adjusted for velocity effects. The length-strength adjustment is made to the latissimus dorsi, erector spinae, and rectus abdominus muscles. The muscle area is calculated from the subjects' torso depth and breadth dimensions and coefficients from Schultz et al. (1982; see Figure 4 and Table 1).

Once all of the other effects have been factored in, an iterative process begins during which the force level is calculated in each muscle. These forces are combined in order to calculate the resulting spinal loading at each point in time (marked by an event for any muscle). That is, whenever an event occurs for one muscle, the activity levels of all muscles are computed at that time and the collective influence on spinal loading is also calculated.

This process begins with the assumption of an initial value of 10 for the gain factor. After setting this initial value, force levels are calculated at each point in time for each muscle. Those values are combined in the equilibrium spinal force and moment equations

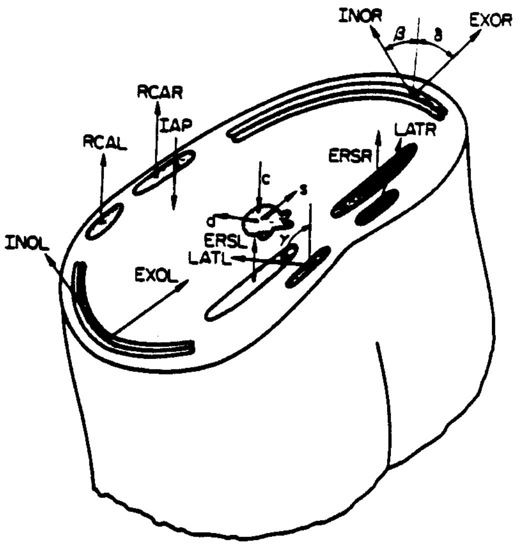


Figure 4. *Cross-sectional view of 10 trunk muscles and their moment arms used in the current model. Angles β, δ, and γ are all assumed to be 45 deg. Refer to Table 1 for coefficients for approximating these areas and moment arm lengths for a given subject. Reproduced with permission from Schultz et al., 1982.*

TABLE 1

Coefficients for Approximating Cross-Sectional Areas and Moment Arms Shown in Figure 4

Muscle	Coefficient for Area	Coefficient for Anteroposterior Moment Arm	Coefficient for Lateral Moment Arm
Rectus abdominus	0.0060	0.54	0.12
Internal oblique	0.0168	0.19	0.45
External oblique	0.0148	0.19	0.45
Erector spinae	0.0389	0.22	0.18
Latissimus dorsi	0.0037	0.28	0.21

Muscle cross-sectional area = coefficient × trunk breadth × trunk depth; anteroposterior moment arm = coefficient × trunk depth; lateral moment arm = coefficient × trunk breadth.
Reproduced with permission from Schultz et al. (1982).

adapted from Schultz and Andersson (1981). The equations used in the current model are listed in Table 2. After this has been done for each event point, a check is performed on the value of the calculated lateral torque. If the average value (area under the curve) is within 5% of the average torque value measured during testing, then the gain is judged to be sufficient and the muscle force and spinal loading values are deemed correct. If the calculated torque is too low, then the gain is increased by a preset increment, the forces are recalculated, and the torques are compared once again. This process continues until the torques are within the acceptable range.

Model output. Once the torques are matched, information on muscle activity pat-

TABLE 2

Equations Used to Calculate Compression, Shear, and Torques in the Current Model

Shear (right-left) = (LATR − LATL) × sin(γ)

Shear (anterior-posterior) = (EXOR + EXOL) × sin(δ) + (INOR + INOL) × sin(β)

Compression = (ERSR + ERSL) + (RCAR + RCAL) + (INOR + INOL) × cos(β) + (EXOR + EXOL) × cos(δ) + (LATR + LATL) × cos(γ)

Torque-X = Ye × (ERSR + ERSL) − Yr × (RCAR + RCAL) + Yl × (LATR + LATL) × cos(γ) − Yi × (INOR + INOL) × cos(β) + Yi × (EXOR + EXOL) × cos(δ)

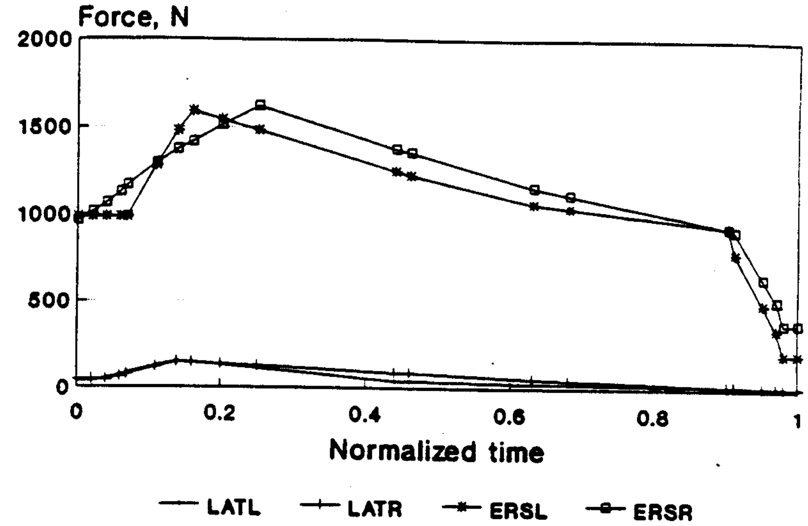
Torque-Y = Xe × (ERSR − ERSL) + Xr × (RCAR − RCAL) + Xl × (LATR − LATL) × cos(γ) + Xi × (INOR − INOL) × cos(β) + Xi × (EXOR − EXOL) × cos(δ)

Torque-Z = Yl × (LATR − LATL) × sin(γ) + Xi × (INOR − INOL) × sin(β) − Xi × (EXOR − EXOL) × sin(δ)

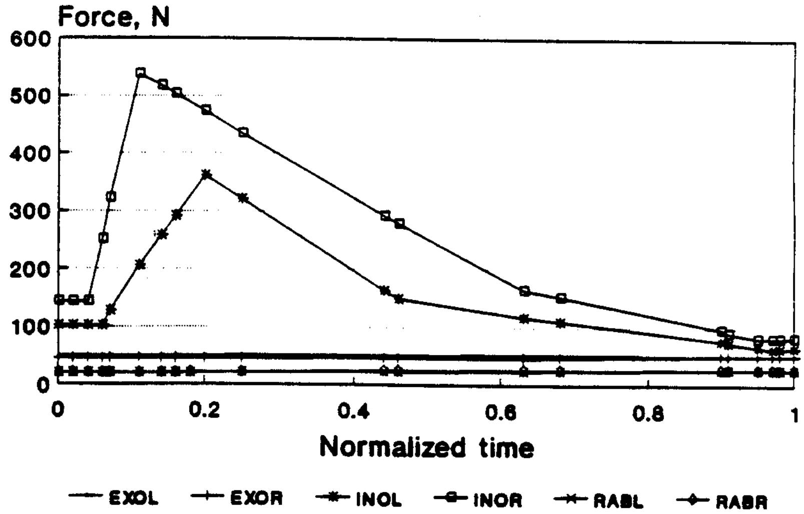
Muscles	Moment Arms: x axis	Moment Arms: y axis
ERSR, ERSL = erector spinae, right and left	Xe	Ye
LATR, LATL = latissimus dorsi, right and left	Xl	Yl
RCAR, RCAL = rectus abdominus, right and left	Xr	Yr
INOR, INOL = internal oblique, right and left	Xi	Yi
EXOR, EXOL = external oblique, right and left	Xl	Yi

Note: Refer to Figure 4 for further clarification.

terns (recruitment order and usage), muscle forces, and spinal loading are written to files for evaluation. These calculated results can then be compared between conditions for a given subject or between subjects for a particular condition. Figures 5a, 5b, and 6 each depicts a typical model output throughout the course of a single exertion. Figures 5a and 5b present muscle activity levels, and spinal loading is shown in Figure 6. When loading



(a)



(b)

Figure 5. *Figures a and b depict model-calculated forces throughout an exertion for the extensor muscles (a) and for the anterior muscles (b). The trunk was in a symmetric posture.*

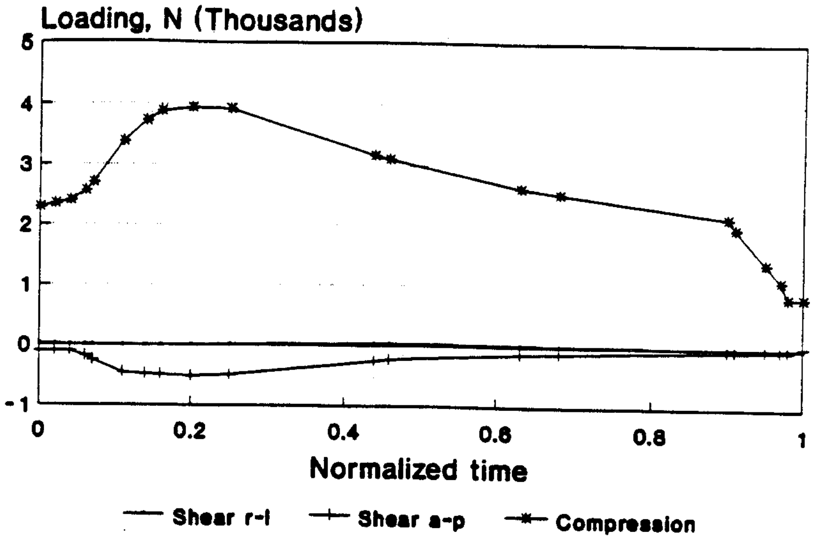


Figure 6. *Spinal loading is depicted over the course of the same exertion as shown in Figure 5.*

levels are compared across conditions, it is possible to study differences caused by changes in externally applied loading and/or changes in trunk speed. These changes of interest may be in the peak or average levels of compression, for example, or any of the other calculated loading parameters.

Individual differences between subjects are revealed when comparisons are made within test conditions across subjects. These can expose differences in muscle recruitment, including differences in the degree of antagonistic activity.

Model Structure Logic

The model structure that governs the estimation of internal force was developed from the six force and moment equations from Schultz and Andersson's (1981) static, internal trunk loading model. However, in the current model these are calculated not once but repeatedly, as the movement progresses from the start to the finish of the isokinetic trunk extension. Reilly and Marras (1989) described an earlier version of the current model, in which they calculated relative trunk loading values for compression and shear, both lateral and anterior, based on a three-point characterization of each of the monitored muscles' EMG records. The version described in this paper varies from that one in a number of ways.

First, a major feature of the current version of the model is the calculation of estimates of actual trunk loading, rather than calculation of relative values, as in the initial model (Reilly and Marras, 1989). Although relative values can reveal differences between loading conditions (e.g., answering a question such as Which condition induces greater peak compression?), values presented in actual units of force can be compared with published results from structural strength tests to determine margins of safety of spinal components under different loading conditions. These muscle force calculations are based on a number of assumptions regarding muscle physiology, which are explained in the next section of the text.

Another modification was the inclusion of a comparison of the calculated torque around the lateral (x) axis (see Figure 4) with that produced by the subject during the trial. The agreement (R^2 value) between these two torques, over the range of the extension, was the metric used to evaluate the goodness of the calculated results for the particular trial.

Mechanical Aspects of Muscle Contraction

One of the major changes included in this version of the model is the addition of several devices to compensate for various factors that are known to affect the EMG signal, including muscle length, velocity of contraction, electrode location, and between-subjects variability.

EMG-force relationship. The electrical activity in a muscle is determined by the number of recruited fibers and the mean frequency of excitation (Bigland-Ritchie, 1981). These factors also determine force generation, so a direct proportionality would be expected. However, many studies by many researchers employing a variety of techniques have come up with linear, parabolic, or even log-log linear relationships between EMG and force (Perry and Bekey, 1981). At lower to midrange force levels, the assumption of linearity holds well enough, though, that this relationship was deemed appropriate and adequate for use in this model. The influence of dynamics on this relationship will be discussed later.

Length-strength relationship. Each muscle has an optimal length at which its force generation capacity peaks. Deviation either way reduces that capacity. This knowledge is incorporated into the model by assuming that each modeled muscle is able to produce a certain maximum amount of tension based on its cross-sectional area (A) and a certain inherent force per unit of area—40 N/cm^2, for example. If the muscle is positioned at its optimal length (Lo), the muscle is assumed to be capable of producing an optimal force ($Fo = Lo \times A$). If, however, the muscle is not at its optimal length, it can produce only some fraction of Fo. The values used to calculate the reduced force are determined by equations derived from several graphs (Brobeck, 1973; Chaffin and Andersson, 1984), as well as some trial and error during model development. Muscle length is assumed to be tied to trunk angle. Perry and Bekey (1981) stated that the proportionality constant relating EMG and force level is dependent on joint position. Asymmetry (twisting or lateral bending) will also affect muscle length, but this is not considered in the model.

Force-velocity relationship. Velocity is known to affect the muscles' tension-producing capabilities during shortening. This is thought to occur because of inefficient coupling at the cross-bridges and, to a minor extent, the fluid viscosity of the muscle. Wilkie (1950) discussed experiments in which he determined that the muscles in the human body follow a fixed force-velocity relationship in contraction against a variety of loads. Bigland and Lippold (1954) demonstrated that during generation of a constant concentric force, higher velocities of shortening were associated with increased levels of integrated EMG activity. The current model incorporates a velocity modulation feature in the form of a ratio. The values that constitute this ratio are based on an extensive data base of EMG records collected in our laboratory during isokinetic extension exertions performed under a variety of velocity and loading conditions.

The velocity modulation ratio is constructed from two values. The numerator is the average normalized EMG response from the laboratory data base for a particular muscle with respect to trunk angle (muscle length), external torque production, and an angular trunk extension velocity of zero. The denominator is the average EMG response, from the same data base, for the same angle and torque conditions but performed at the speed at which the trial of interest was conducted. This ratio is multiplied by the normalized EMG value for the subject in order to eliminate the velocity effect. Equation 2 specifies this velocity modulation ratio:

$$\textit{Velocity modulated } EMG(i,j,k,0) = \textit{Subject } EMG(i,j,k,l) \times \frac{\textit{Average } EMG(i,j,k,0)}{\textit{Average } EMG(i,j,k,l)}, \quad (2)$$

where i = muscle of interest, j = trunk angle, k = torque exerted by subject, and l = angular velocity of trunk (0 to 30 deg/s). This reduces the likelihood of overestimating the amount of force from a given muscle when the EMG signal is elevated because of the velocity of the contraction.

EMG normalization. In order to compare EMG levels (translated into units of force in the model) across trials and subjects, it was necessary to normalize the EMG values. This procedure compensates for subject differences as well as muscle differences. Three maximum EMG values for each muscle of each subject were recorded during the performance of static maximum exertions at three equidistant trunk flexion angles. This technique utilizes the concept of the window-of-view of the surface electrode and how that view changes as the muscle length is altered. A greater or lesser number of motor units may be viewed depending on the muscle's orientation. Dividing an EMG value by the maximum value recorded for that muscle aids in the understanding of what the original EMG value means—that is, how hard the muscle is working at that angle of orientation. This technique permits equitable comparisons of activity levels throughout a movement.

Gain. This is a factor that accounts for the maximum force per unit of area in the muscles. Pope, Andersson, Broman, Svensson, and Zetterberg (1986) incorporated a similar factor, which, they explained, related muscle stress to the EMG signal and was a function of such factors as the distance of the electrode to the muscle and skin preparation. In the current model the gain factor also compensates for deficiencies in the other modulation techniques. That is, if all of the muscle's mechanical aspects were accounted for through the specific modulation factors previously listed, the gain would be expected to be the same for every trial for a given subject.

DISCUSSION

This paper has described a model that can estimate spine compression, shear, and torsion based on subject anthropometric information, EMG activities of the trunk muscles, and trunk kinematic information. This is not a general-purpose model but, rather, is intended for use under laboratory conditions. The model is intended to interface with common laboratory instruments (EMG, dynamometers, etc.) so that the influence of motion-related biomechanical factors can be assessed. Under these conditions it can be a valuable tool for understanding how the spine is loaded as a result of certain occupational factors.

There are several advantages of such a model. First, many ergonomic and occupational biomechanics studies have collected trunk EMG data and reported differences in activities attributable to occupational conditions. However, there is often little quantitative interpretation of these EMG data in terms of the effects of these muscle activities on spine loading. This model provides a method to consider the data synergistically, so that the collective influence of the muscle activities on spine loading can be appreciated.

Second, this model provides a means to identify peak spine loadings caused by dynamic experimental conditions. Many predictive models report average spine loading, often over prolonged periods of exertion. However, this model would permit one to evaluate spine loading at points in time throughout the exertion. Peak spine forces could be pinpointed and later compared with spine tolerance limits, thereby yielding a much more accurate assessment of the risk of injury attributable to workplace design.

Another advantage of this approach is that it provides information as to how individuals create loading on the spine during trunk motion when various torques are imposed on the trunk. Factors that are necessary for proper spine load evaluation are included in this analysis and include an understanding of coactivity of the trunk musculature as well as an understanding of effects of three-dimensional motions of the trunk. This model also has the advantage that it has acceptable accuracy and yet is not overly complex.

This model could be used to advance the state of the art in manual materials handling analysis. One could vary the model parameters to simulate the trunk motion, trunk torque, and trunk position that are observed in the workplace and to determine those workplace ergonomic factors that increase the risk of excessive spine loading. The model could also be used to study the systematic coactivity patterns of the trunk musculature. This would permit one to use the model as a tool to develop more accurate predictive models of the trunk under occupational lifting conditions.

Future improvements of this model could include other parameters of dynamic motion. For example, it could be tested or adjusted to respond to eccentric or muscle-lengthening types of exertions. This could be accomplished by including a muscle force lengthening component in the gain factor. Similarly, the trunk acceleration component of trunk motion could also be investigated by adjusting the gain factor or by adding an EMG acceleration modulation component to the model. It is expected that a sufficient number of trunk motion components could eventually be investigated, in this piecewise manner, so that an accurate understanding of spine loading under true dynamic lifting conditions could be gained.

Finally, this model has been tested and validated under the experimental conditions described in this paper. A companion article (Marras and Sommerich, 1991 [this issue]) describes the experimental conditions, model performance, and accuracy of the model.

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