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# The Development of an EMG-Assisted Model to Assess Spine Loading During Whole-Body Free-Dynamic Lifting

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**Summary:** Low back disorders (LBDs) are the most common and costly occupationally-related compensable conditions facing employers today. Over the years several biomechanical assessment models have been developed that intended to assess the load profile imposed upon the spine during lifting and, thus, intended to facilitate the control of LBD risk in the workplace. Many of these biomechanical models have evolved based upon assumptions about how the trunk musculature respond to loads imposed upon the body during lifting. However, few of these models have been able to accurately predict the co-contraction of the trunk musculature which has been shown to have a major influence on the development of spinal loads. Thus, our understanding of how the spine is loaded under realistic dynamic lifting conditions has been deficient. A biologically-assisted or EMG-assisted model has been developed in our laboratory over the past 15 years which endeavours to overcome these traditional problems. The model has been assessed in the sagittal, coronal, and torsional planes of the body. The model development and performance will be reviewed as well as the benefits for controlling occupationally-related LBDs. © 1997 Elsevier Science Ltd. All rights reserved

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## INTRODUCTION

Occupationally-related low back disorders (LBDs) are currently and have been the leading cause of lost work days as well as the most costly occupational safety and health problem facing industry today. It is well known that most occupationally-related LBD risk is associated with manual materials handling (MMH) tasks. However, a major limitation in controlling the incidence of occupationally-related LBDs has been the inability to accurately assess the loading that occurs on the lumbar spine (the most common site of injury) during realistic, whole-body free-dynamic MMH conditions. An accurate evaluation of spine loading is necessary so that the loads imposed upon the lumbar spine can be compared with tolerance limits of the spine derived through cadaver studies as well as finite element models of the spine.

Historically, most biomechanical models used in ergonomics have made assumptions about the conditions under which work is performed. These assumptions limit their applicability to realistic dynamic loading conditions. Models were originally static and did not consider the effects of motion. Latter models were dynamic but did not consider

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the effects of the collective influence of the trunk structures upon the loading of the spine[5]. These models were deficient in that they were not able to explain how the trunk muscles work collectively (co-contraction) to support the external load and simultaneously impose loadings upon the spine. Ignoring co-contraction is undesirable in that the estimates of spine compression could be under-predicted by 45% and estimates of spine shear loading could be under-estimated by as much as 70%[7]. In addition, such models are problematic in that they are not able to account for the variability in muscle recruitment and the subsequent variability in spine loading that occurs from trial to trial during repetitive lifting bouts. Thus, neither of the previous classes of models were able to accurately assess the risk of joint loading during dynamic workplace circumstances.

A key element of work, trunk motion, could be beneficial as well as detrimental. In some circumstances it is believed that ballistic motions could actually minimize energy transfer through the body[4] thus reducing or at least changing the form of the joint loading. Therefore, it is imperative that one understands how the musculature behave under dynamic loading conditions. The significance of accounting for this trunk muscle co-contraction has been recognized by many modellers. In order to include such activities some have used optimization techniques[1,26] to estimate muscle activities. However, due to the indeterminacy of the muscle system these models were not able to account for the co-contraction that occurs among the muscles surrounding a joint during dynamic motion.

In order to address these limitations of previous efforts, a free-dynamic three-dimensional biodynamic model of the trunk has been under development in the Biodynamics Laboratory at the Ohio State University for the past 15 years. The model accounts for the collective co-contraction influence of 10 trunk muscles upon the three-dimensional loading of the spine. We have developed this model to the point where we can accurately predict three-dimensional loading of the spine under free-dynamic forward bending, lateral bending, and twisting of the lumbar spine.

## MODEL DEVELOPMENT

Efforts to describe the activation of the trunk musculature during dynamic trunk loading began with the work of Marras[12]. Using this data, modelling efforts began with the development of descriptive network models of the time event changes that occur under various motion conditions[13,17]. The original research investigated the time sequence activity of 10 trunk muscles and IAP as subjects produced trunk extension motions (lifting motions) in the sagittal plane. Based upon these observations, descriptive network models of dynamic exertions were developed[17]. These networks described the sequence of trunk structure components that developed under various velocity conditions. This analysis indicated that there are indeed velocity-dependent sequence changes that occur when the trunk moves at different rates of speed. The complexity of this timing is shown in Figure 1. This type of timing information becomes important in the development of dynamic models since dynamic models need to assess the impulse load imposed upon the spine during a work situation. Since impulse information is time dependent, the only way to relate that information is through time sequence information. It is believed that this impulse information represents a “worst case” assessment of spine loading which is imperative for cumulative trauma assessment. These authors also used this information regarding muscle activities to create a crude EMG-driven simulation model of the spine called SIMULIFT[25].

Since the EMG signals represent a continuous measure of muscle activity, the EMG signal could be used as a basis for determining force generation history of each muscle surrounding a joint. This logic has become the basis for some of the new EMG-assisted models which have been introduced recently for the assessment of dynamic motion. One of the early models to use this information was developed by McGill and Norman[20,21]. This model used the EMG signals from six muscles to estimate muscle force in 20 muscles and predicted the contribution of eight ligaments and fascia. This information was used to predict spine loading during sagittally symmetric dynamic lifting performed by three subjects. The model used theoretical relationships to adjust for muscle force based upon muscle length, cross-sectional area, and velocity.

Following similar model logic, Marras and Sommerich[18,19] developed a 10-muscle EMG-assisted model which was capable of predicting spine forces under asymmetric dynamic bending conditions. The rationale for our model development has been to only make the model as complex as necessary to accurately represent external moment and, thus, spine load. In addition, the goal was to only include

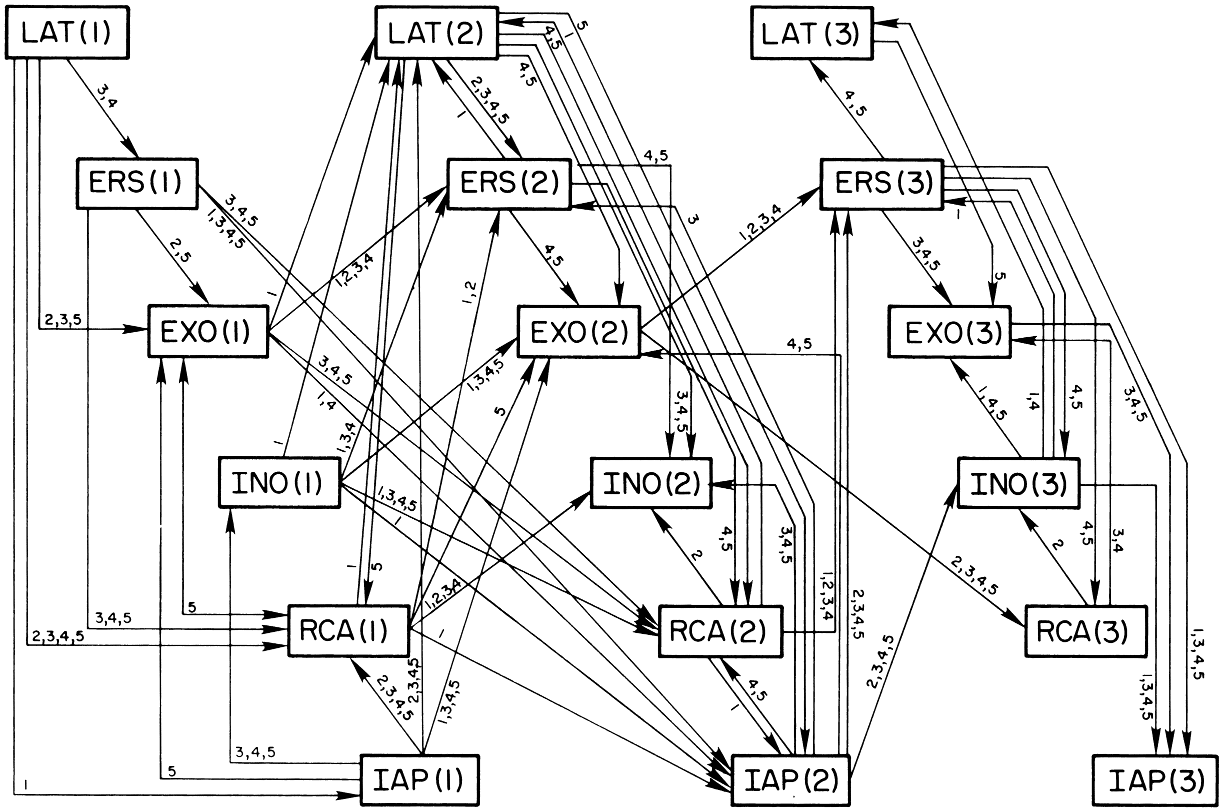


**FIG. 1.** A network representation of trunk muscle initial activation points (1), peak activity points (2) and activity termination points (3) for the erector spinae muscle (ERS), latissimus dorsi (LAT), internal oblique (INO), external oblique (EXO), the rectus abdominus (RCA) muscles and intra-abdominal pressure (IAP). Numbers at each line represent the velocity condition in terms of percentage of maximum voluntary velocity where 1 = static (or 0% of maximum velocity), 2 = 25% of maximum velocity, 3 = 50% of maximum velocity, 4 = 75% of maximum velocity, and 5 = 100% of maximum velocity.

muscles that can be documented via direct EMG measurement so that muscle activity assumptions could be avoided. Geometrically, this model assumed that one could represent the trunk mechanically via a description of the transverse cutting plane passed through the lumbar spine[26]. This model used both theoretical relationships as well as empirical data from 100 subjects to adjust muscle force based upon EMG. EMG activity from 10 muscles was used to predict muscle forces based upon muscle length, EMG pick-up volume, velocity, gain, and cross-sectional area. For simplicity purposes, this model described EMG activity via linear geometric representations of the EMG time history signal. An example of one of these figures is shown in Figure 2. This signal was then adjusted for muscle length and velocity. This model predicted spine forces as well as trunk torque production. Measured trunk torque was compared to predicted trunk torque and used as a validation measure. Two validation experiments[6,8] involving over 30 subjects have shown that

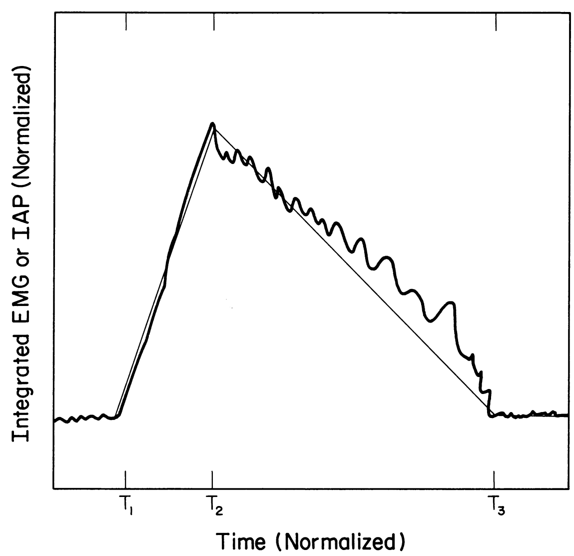


**FIG. 2.** Linear geometric approximation of muscle activity used for computational efficiency in early EMG-assisted models.

the model accounts for an average of over 85 percent of the torque variability. As asymmetry and velocity increased significant increases in spine compression were found. Similar trade-offs between velocity and loading occurred with shear forces.

## General Description

Significant improvements have been incorporated into our EMG-assisted free-dynamic biomechanical model over the past several years[8,14,15]. Compared to former EMG-assisted models this model is unique because multi-dimensional, trunk moments and spinal loads are determined from dynamic muscle force vectors and moment arms. Our current EMG-assisted free-dynamic lifting model employs 10 muscle equivalent vectors to approximate trunk anatomy and mechanics. The muscle equivalents are treated as simple cables tensioned between points of origin and insertion. Active muscle forces are assumed to be an adequate description of trunk mechanics without consideration of passive muscle, ligament, or disc forces. At trunk extreme flexion-extension angles passive forces may become significant but within the design range of 45 degrees flexion to vertical the trunk moments may by active muscle forces. Our field studies[16] have shown that little industrial lifting requires trunk bending beyond 45 degrees.

Muscle fibres sampled by EMG surface electrodes are assumed to be representative of, and linearly related to the net muscle force. Lippold[11], Moritani and DeVries[23], and Yoo *et al.*[28] demonstrated linear relationships between surface EMG activity and voluntary isometric joint torque. Conversely, Zuniga and Simons[29], Vredenbregt and Rau[23], and Komi and Viitasalo[10] measured EMG proportional to the square of the isometric joint torque. It has also been shown that a single muscle can produce a linear or non-linear EMG-force relationship at the tendon depending upon the function performed[27]. In order to simplify this issue, our model employed a linear EMG-force relationship. In addition, we assume this relationship is valid for those motions wherein the time delay between the onset of myoelectric activity and muscular contractile force is minimal, i.e. smooth lifting motions.

The model employs EMG and kinematic input to determine the dynamic, relative, muscle, force vectors of the 10 modelled trunk muscles. Relative force vectors are scaled by a gain factor computed from input kinetic data. Predicted, multi-dimensional, dynamic, trunk moments and spinal loads are computed from muscle force vectors and muscle moment arms determined from subject anthropometry and kinematics.

## Trunk Mechanics

Whereas previous models assumed muscle vector directions were constant in space, the free-dynamic model allows each muscle orientation, length, and velocity to vary with the lifting motion and position of the trunk. Muscle origins were assigned a three-dimensional location relative to the spinal axis, co-planar with the iliac crest. Muscle insertions are located co-planar with the 12th rib. Muscle forces were represented as vector quantities between their two endpoints. In essence, the mechanics of this model can be visualized as two "plates" that can be allowed to move relative to one another[14]. These plates represent the attachment point of the muscle in the pelvis and thorax. The muscles were represented by vectors between these plates that change their orientation as the trunk moves dynamically (Figure 3). Using this approach, muscle orientations and lengths change throughout a movement, thus, accounting for a muscle's changing mechanical advantage throughout the task. In addition, such a representation affords the opportunity to change the relative angles between the iliac and thoracic planes. Thus, pelvic tilt or asymmetric bends of the trunk could be simulated. Muscle origins and insertions were dynamically located via Euler rotation of the

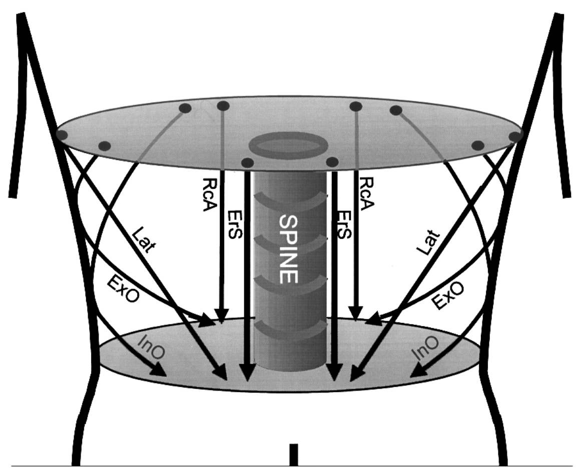


**FIG. 3.** Vector representation of the trunk used in the EMG-assisted model. This representation facilitates the adjustment of muscle orientation and the muscle length–strength relationship during trunk motion. Muscle vectors are represented by curvilinear lines to enhance interpretation in the figure.

anatomically defined, three-dimensional coordinates relative to the measured trunk motion. Muscle vector directions, lengths and velocities were continuously determined from the instantaneous positions and motions of the muscle endpoints. Time- and position-dependent force vectors significantly affect the predicted trunk moments and forces generated by the musculature, by allowing the vector direction to move throughout an exertion.

Spinal loading (compression, right-lateral shear, and anterior shear forces) were calculated from the vector sum of validated muscle forces. Muscle generated moments about the spinal axis were predicted from the sum vector products combining dynamic tensile forces of each muscle, j, and respective moment arms.

$$\vec{M} = \sum_{j} \vec{r}_j \times \vec{F}_j \quad (1)$$

Spinal compression, right-lateral shear force, and anterior shear force were displayed as a function of time. Measured and predicted values of the trunk moments, as well as predicted compression, anterior shear, and lateral shear forces were written to a file for post-modelling analysis. The task gain and correlation between measured and predicted moment profiles were recorded for model performance evaluation.

Measured and predicted trunk moments were compared and must agree if the model is correctly simulating trunk mechanics. Statistical correlations between predicted and measured moment profiles ($R^2$) serves as the measure of model performance and indicates how well the model accounted for the variability in the motion moment. A high correlation implies the model generates an accurate simulation of dynamic spinal loading. Of course, there is no means by which one could actually measure spine loading *in vivo*. However, we feel that if the model could accurately predict applied moments about the spine then the predictions of spine loadings should also be reasonable.

## Muscle Force Estimates in the Model

Relative muscle contractile force magnitudes were computed from normalized EMG, modulated to account for muscle length and velocity, and scaled by muscle cross-sectional areas. Relative muscle activities were multiplied by an appropriate muscle force per unit cross-sectional area, i.e. gain, determined from the solution of dynamic equilibrium. Previous EMG-assisted models employed muscle areas representative of the cross-sections found at the lumbosacral transverse plane. The current model recognizes that the force generating capacity of the latissimus dorsi was poorly represented by the small slips of muscle that pass through the lower lumbar levels. Consequently, the maximum cross-sectional area of the latissimus dorsi, found near T5[22], was employed in the model to scale the force-generating capacity of that muscle. Similar maximum cross-sectional areas were used to represent the maximum force generated by the other trunk muscles.

The tensile force generated by each muscle, j, was described (equation 2) by the product of normalized EMG, muscle cross-sectional area, a gain factor representing muscle force per unit area, and modulation factors describing EMG and force behaviour as a function of the length f($\mathrm{Length}_j$), and velocity, f($\mathrm{Vel}_j$) of muscle j.

$$Force_j = Gain\frac{EMG_j(t)}{EMG_{Max\,j}}Area_j f(Vel_j) f(Length_j) \quad (2)$$

EMG data are normalized relative to myoelectric maxima from each muscle. This was necessary to remove possible analytical errors related to electrode placement, skin abrasion, flesh resistance, muscle fibre density, and electronic channel differences.

Relative myoelectric activities were multiplied by a unitless function of length, f(Length), to account for the relation between tensile force and muscle length. The modulation factor incorporates physiologic length–strength relations and artifact due to variation in the myoelectric potential density picked up by the surface electrode. The functional coefficients were determined by minimizing the average variation in predicted gain as a function of length. The length modulation factor (equation 3) employs the instantaneous length of muscle, j, determined from the anthropometry coefficients and kinematic inputs.

$$f(Length_j) = -3.2 + 10.2Length_j - 10.4Length_j^2 + 4.6Length_j^3 \quad (3)$$

The empirically determined coefficients agree with the expanded form of the length–modulation factor proposed by McGill and Norman[21].

Relative muscle activity was also multiplied by a unitless function of contractile velocity, f(Vel), to account for the physiologic force–velocity relation and associated EMG artifact. Bigland and Lippold[2] demonstrated increased muscle contraction velocity yields increased myoelectric activity without a concomitant increase in force output. The modulation coefficients were computed by minimizing the average variation of gain predicted by the model as a function of velocity. The velocity modulation factor (equation 4) included the time-dependent, contraction velocity of each muscle, j, determined from anthropometry and kinematic data.

$$f(Vel_j) = 1.2 - 0.99Vel_j + 0.72Vel_j^2 \qquad (4)$$

The coefficients for the modulation factor approximated the theoretical force–velocity relationship represented in the Hill[9] equation, and are appropriate for smooth, non-ballistic muscle contraction.

Normalized and modulated EMG data were multiplied by their respective muscle cross-sectional areas to account for the relative force generating capacity of each muscle. It has been demonstrated[3] that maximum muscle force is directly related to cross-sectional area for fusiform muscles. Therefore, scaling the EMG by muscle area provided larger muscles with greater modeled force-generating capacity.

Gain, i.e. muscle force per unit area, was computed by comparing muscle-generated trunk moments with measured applied moments about the lumbosacral junction. To satisfy the equations of dynamic equilibrium, the muscle generated extension moment must equal the measured moment. Gain was appropriately and automatically adjusted to satisfy this condition. To be physiologically valid, the predicted gain level must fall within the range of 30 to 100 N/cm$^2$[24]. Muscle force per unit area is highly variable between subjects, based on subject conditioning and natural ability. On the other hand, gain predicted for a given subject must be constant throughout each of the experimental trials. Examination of the gain value and its within-subject variability provided a means for testing model validity.

## Model Input

Input data required by the model includes time-domain EMG, exertion kinetics, and kinematics. Maximum exertion EMG levels and subject anthropometry were also employed to calibrate and format the dynamic data suitable for use in the model mechanics. The cross-sectional area of each muscle was computed from regression equations based on the subject's trunk depth and breadth.

Voluntarily applied external kinetics, including gravitational moments and acceleration effects on trunk mass were dynamically measured by a force plate and pelvic stabilization system. Translation of force plate mechanics was performed to compute three-dimensional force and moments about the lumbosacral spine. The pelvic stabilization system permits free-dynamic motion above the pelvis. More recently this system has been modified to permit free dynamic motion of the whole body. This was accomplished by mathematically correcting for the position of the pelvis relative to the force plate.

Trunk velocity was computed from dynamic measures of trunk flexion, twist, and lateral angles collected from a lumbar motion goniometer (lumbar motion monitor or LMM)[16]. The LMM offers no significant restriction to trunk mobility, while accurately recording trunk kinematics. Collected kinematic data were used in the lifting model to: i) describe the trunk motion as a function of time, ii) determine the muscle force and moment vector directions, and iii) modulate muscle EMG values to account for muscle length and velocity artifact.

In the current form of the model, EMG data were collected from the right and left latissimus dorsi, erector spinae, rectus abdominus, internal abdominal obliques, and external abdominal obliques. The time-domain myoelectric data represent muscle activity and are used to calculate relative muscle force. EMG signals were collected from circular Ag/Ag Cl 4 mm diameter surface electrodes. The signals were pre-amplified at the muscle site, high-pass filtered at 30 Hz, and low-pass filtered at 1 000 Hz in the main amplifier. Then the signal was full-wave rectified and processed (smoothed) via a 20 ms sliding window hardware filter. Maximum and resting EMG values were collected from flexion and extension exertions to normalize the dynamic EMG signals.

All dynamic data, including kinetics, kinematics, and EMG, were smoothed via a Hanning weighted time-domain filter within the model. Smoothing the data was necessary to remove digitizing noise and artifact from differentiation and calibration routines. The model has been developed in a Windows™ environment which permits spine loading and biomechanical activity to be assessed relative to lifting activity. An example of the Windows model is shown in Figure 4.

## Model Performance and Validation

The free-dynamic model has been tested in three separate experiments. The first experiment was

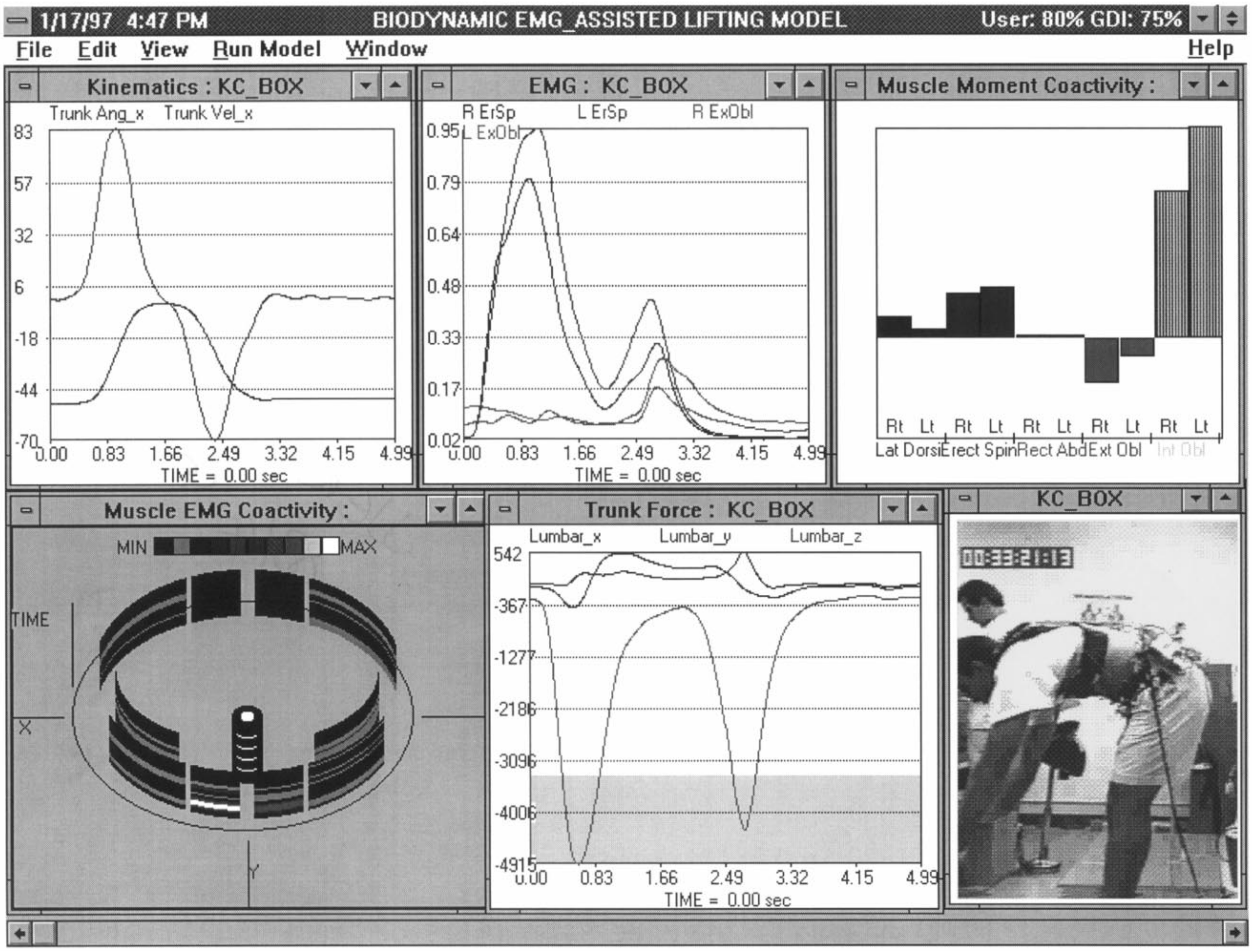


**FIG. 4.** An example of the EMG-assisted model used in a Windows environment. This environment assists in not only understanding the biomechanical function during a lift but also permits one to associate the biomechanical behaviour with the video representing the lift of interest.

designed to test the ability of the model to assess symmetric and asymmetric lifting motions under isokinetic compared to free-dynamic conditions[8]. The second experiment was intended to evaluate the ability of the model to assess spine loading during trunk twisting[14]. The last experiment was used to assess spine loading during pure lateral bending motions[15].

In the first study[8], the EMG-assisted model was exercised and results generated from 703 separate lifting exertions designed to test its validity under free-dynamic conditions and to compare its performance with previous models of trunk mechanics. Ten subjects lifted loads of 0, 40, and 80 lbs at isokinetic trunk angular velocities (30, 60, and 90 deg/s) as well as free dynamically (slow, medium, and fast) lift rates. Subject gain values averaged over all free-dynamic exertions was 47.4 ± 12.1 N/cm$^2$ and fell within the physiologically acceptable range. Because subject's muscle strength per unit area can not change from one exertion to the next, a subject's gain value must remain constant. Although gain changed significantly ($P < 0.01$) between subjects, the values did not vary significantly within subjects. Thus, the model predicted a muscle force per unit cross-sectional area which was physiologically-valid. Distributions of squared correlation coefficients that indicated the association between measured and predicted external trunk moments were derived from the dynamic lifting trials. They illustrate that over 88.6% of the trials performed with an $R^2$ greater than 0.80, and 68% performed with an $R^2$ greater than 0.90. Statistical analyses consisted of ANOVA evaluations of the differences in gain, average absolute error, and $R^2$ as a function of the experimental conditions, trials, and subjects. These evaluations

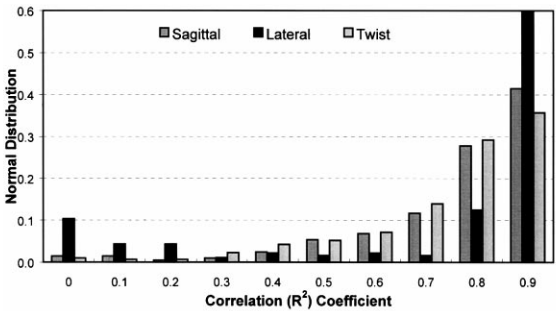


**FIG. 5.** $R^2$ distribution indicating the relationship between measured moment and predicted moment for forward bending motions (sagittal), twisting motions (twist), and lateral bending motions (lateral).

demonstrated that the model performed well independent of the type of lifting exertion. There was no statistically significant difference between the isokinetic ($R^2 = 0.87$) and free-dynamic exertions ($R^2 = 0.89$). Figure 5 shows the $R^2$ distribution for this data set. The model performed well at all dynamic velocities.

Model predictions of compressive, lateral shear, and anterior shear loading agree with trends cited in previous studies[18,19], however, we believe our results more accurately represent loads that occur from primary and co-active muscle activity during realistic lifting tasks. Spinal loads generated throughout dynamic lifting exertions increased as a function of trunk asymmetry and lifting velocity. The impact of dynamic loading compared to static loading can be appreciated in Figure 6. This figure shows that both compression and shear increase as the velocity of motion increases.

In the second study[14], 12 males, 21 to 31 years-of-age participated in an experiment involving torsional moment, torsional direction, twisting position, and twisting velocity. Subjects were asked to twist under static loading conditions as well as at velocities of 10 and 20 deg/s. Industrial studies have indicated that these slow twisting velocities increase risk of LBD on the job[16]. Spine loading, as predicted by the EMG-assisted model, was significantly affected by many of the variables manipulated in this experiment. Statistical comparisons indicates that relative spinal loads (per unit of torsional moment) changed as a function of exertion level, direction of the applied twisting torque, and twisting velocity. Any level of twist velocity substantially increases relative spine forces in all three cardinal planes compared to isometric exertions (Figure 7). Predicted values of the calibrated subject muscle gains indicated that the model was also valid for twisting exertions. The model performed well during both static and dynamic exertions. The squared correlation coefficients ($R^2$) between measured and predicted torsional moments from the 320 trials indicated that the average value was 0.80 and is also shown in Figure 5. Predicted gain was 35 N/cm$^2$.

The third study[15] also employed 12 subjects between the ages of 24 and 33. These subjects were asked to support a lateral moment both statically (while positioned in an upright position or laterally bent to the right or left at 15 degrees) as well as while they were moving isokinetically at 15, 30, or 45 deg/s. As with the sagittal bending and twisting study, the model performed well. Average gain values were below 65 N/cm$^2$ and the average $R^2$ was 0.91 (Figure 5). As indicated in Figure 8, lateral shear forces increased quickly as lateral velocity increased.

## DISCUSSION AND CONCLUSIONS

These efforts have demonstrated the steps involved in the development of a free-dynamic three-dimensional model of the spine that is capable of accurately assessing spine loadings during trunk

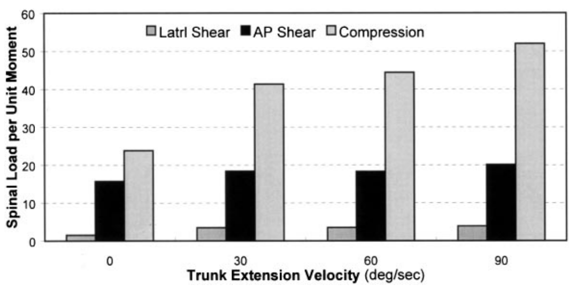


**FIG. 6.** Spine compression and shear (in load per unit moment) shown as a function of trunk extension velocity.

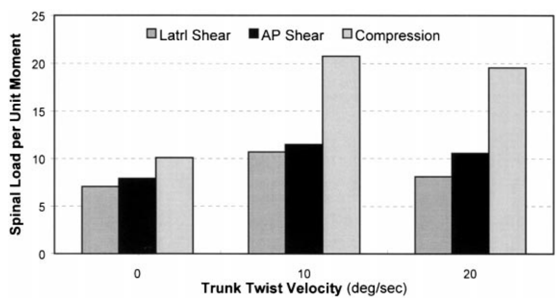


**FIG. 7.** Spine compression and shear (in load per unit moment) shown as a function of twisting velocity. Note that co-contraction results in increased compression even at very low velocity.

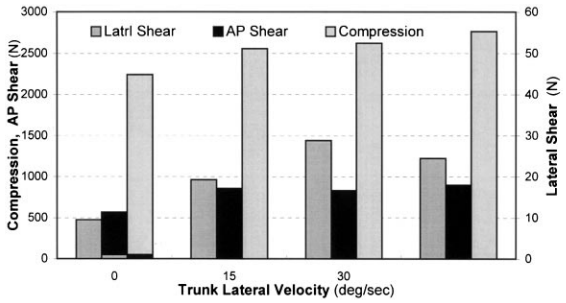


**FIG. 8.** Spine compression and shear shown as a function of lateral bending velocity.

motion during manual materials handling under laboratory conditions. We have developed this model based upon the logic that the model need not be unnecessarily complex in order to assess the loads imposed on the lumbar spine during dynamic three-dimensional trunk lifting motions. Since the objective of the model has been to describe the gross loads on the lumbar spine, we feel that a model of the complexity described here is adequate and useful for ergonomic purposes. It is also acknowledged that more elaborate models are necessary for other purposes. For example, if one wishes to assess the load imposed upon a specific tissue, a much more complex model would be necessary. However, along with this complexity one must make numerous assumptions which detract from the appeal of a biologically-assisted model.

The results of these studies have been of practical use to those interested in the dynamics of workplace lifting. Industry-based studies have clearly indicated how increases in work-related trunk bending, trunk lateral velocity, and trunk twisting velocity can significantly increase the risk of low back disorder[16]. Collectively, the model results from the studies described here indicate that the common component in all of these occupational trunk motions is increased co-contraction of the trunk muscles. We have demonstrated that it is imperative to assess this co-contraction during work or spine loading can be underpredicted significantly[7]. Thus, we have gained a new understanding for how and why dynamic trunk motions increase the risk of occupationally-related low back disorders. These studies also suggest that model strategies that fail to accurately account for this co-contraction under realistic dynamic loading conditions should be used with caution since they are likely to misinterpret the loading of the spine. Until better muscle recruitment prediction models and methods are developed EMG-assisted models appear to be the most accurate means to assess realistic spine loading. Of course the price of using such models lies in the instrumentation requirements which are often impractical for some workplaces. Thus, these work situations are often simulated in the laboratory.

Future embellishments of the model will endeavour to incorporate a better understanding of passive tissue mechanics on spine loading as well as make the model more usable during activities at the workplace.

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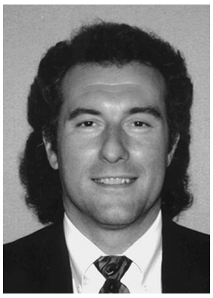

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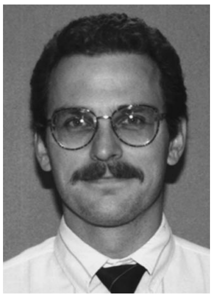

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