J. Biomechanics Vol. 26, No. 12, pp. 1429–1438, 1993.
Printed in Great Britain

0021-9290/93 $6.00+.00

AN EMG-ASSISTED MODEL OF LOADS ON THE LUMBAR SPINE DURING ASYMMETRIC TRUNK EXTENSIONS

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Abstract—An EMG-assisted, low-back, lifting model is presented which simulates spinal loading as a function of dynamic, asymmetric, lifting exertions. The purpose of this study has been to develop a model which overcomes the limitations of previous models including static or isokinetic mechanics, inaccurate predictions of muscle coactivity, static interpretation of myoelectric activity, and physiologically unrealistic or variable muscle force per unit area. The present model predicts individual muscle forces from processed EMG data, normalized as a function of trunk angle and asymmetry, and modified to account for muscle length and velocity artifacts. The normalized EMGs are combined with muscle cross-sectional area and intrinsic strength capacity as determined on a per subject basis, to represent tensile force amplitudes. Dynamic internal and external force vectors are employed to predict trunk moments, spinal compression, lateral and anterior shear forces. Data from 20 subjects performing a total of 2160 exertions showed good agreement between predicted and measured values under all trunk angle, asymmetry, velocity, and acceleration conditions. The design represents a significant step toward accurate, fully dynamic modeling of the low-back in multiple dimensions. The benefits of such a model are the insights provided into the effects of motion induced, muscle co-activity on spinal loading in multiple dimensions.

INTRODUCTION

Computer models of lumbro-sacral biomechanics during lifting exertions have progressed from static, two-dimensional analysis (Chaffin, 1969; Chaffin and Baker, 1970; Schultz and Anderson, 1981), to more recent attempts at understanding dynamic, three-dimensional, stresses on the spine (Marras and Sommerich, 1991a; McGill, 1991, 1992; Pope *et al.*, 1986). These models attempt to represent accurately and realistically the mechanical loading and behavior of the low-back, while refraining from as much unnecessary complexity as possible. Evidence (Frievalds *et al.*, 1984; Goel *et al.*, 1991; Marras and Sommerich, 1991b; McGill and Norman, 1985) has shown that static lifting models underpredict dynamic, trunk, extension moment and spinal loading by as much as 30 to 40%, whereas isokinetic models overpredict peak, dynamic lifting moment by an average of 25% (McGill and Norman, 1985). Literature demonstrates that multidimensional, coupled, dynamic motion is the major risk factor for injury in manual materials handling (Bigos *et al.*, 1986; Marras *et al.*, 1993; Shirazi-Adl, 1989; Shirazi-Adl *et al.*, 1986; U.S. Dept. Labor, 1982). Therefore, lifting models must be extended to accurately predict spinal loading experienced during three-dimensional, fully dynamic motion.

Significant muscle coactivity has been measured as function of exertion, load, trunk position, velocity, and acceleration (Gracovetsky *et al.*, 1985; Marras *et al.*, 1984; Marras and Mirka, 1992a, b; McGill and Norman, 1988; Sudhakar, 1990; Zetterburg *et al.*, 1987). Thus, spinal loading predicted from a single equivalent trunk muscle (Chaffin, 1969; Chaffin and Baker, 1970) may be inaccurate (Hof and Van Den Berg, 1977). Optimization techniques used to estimate muscle forces from the indeterminate system of equations resulting from multiple muscle models (Goel *et al.*, 1991; Gracovetsky and Farfan, 1986; Jager and Luttmann, 1989; Schultz and Anderson, 1981), typically drive antagonistic activity to unrealistic levels (Hardt, 1978; Marras, 1988). Assumption of insignificant coactivity often ensures underestimation of spinal compression.

Dynamic models have been developed using electromyographic (EMG) measures to estimate the force in trunk muscles. The advantages of these models are that (1) they are not limited by the constraints of optimization objective functions (2) they account for muscle coactivation forces via measurement (3) they typically use predicted, physiologic coefficients for instantaneous validity checking. Myoelectric, bilateral symmetry assumptions have been employed (McGill and Norman, 1986), but disregard risks from induced lateral shear and torsion at the base of the spine (Shirazi-Adl *et al.*, 1986; Shirazi-Adl, 1989). Predictions of sagittally symmetric and asymmetric lifting moments and associated spinal loads have been achieved via EMG data from five, left–right pairs of trunk muscles (Marras and Sommerich, 1991a; Reilly and Marras, 1989). Good correlations were produced from these models by approximating dynamic data profiles as straight line segments, but this reduces the true power of the EMG-assisted model by artificially representing the measured muscle activity. Relative muscle force in these models is determined from EMG activity normalized to a maximum, but maximal and submaximal EMG activity changes significantly with trunk angle, isokinetic velocity, and acceleration (Marras *et al.*, 1984, 1986; Marras and Mirka, 1990, 1992a,b). Previous EMG-assisted models employing a single, constant, maximum value for EMG processing

Received in final form 26 *February* 1993.

(McGill and Norman, 1986; Reilly and Marras, 1989) may incorrectly represent relative muscle forces occurring under dynamic lifting conditions. To model accurately low-back mechanics, dynamic effects on myoelectric activity and muscle force must be included by normalizing the EMG input as a function of both trunk bending angle and asymmetry, and modifying for length and velocity artifact (Marras and Sommerich, 1991a; Marras and Mirka, 1992a). EMG-assisted models require a gain value representing the subject's muscle strength per unit area. This value is either input as an assumed value (McGill, 1992; McGill and Hoodless, 1990; McGill and Norman, 1986) or derived from the model mechanics (Marras and Sommerich, 1991a; Reilly and Marras, 1989). Clearly, the muscle force per unit area may be different for each subject, but must not change from trial to trial. Some previous EMG-assisted models treated the gain as an error term by allowing it to vary from trial to trial. No EMG-assisted model has attempted to compute and control the gain as a subject-dependent constant. Published twisting models (McGill, 1991; Pope *et al.*, 1986) have suffered from physiologically unrealistic gain values (McGill and Norman, 1987; Reid and Costigan, 1987; Weis-Fogh and Alexander, 1977).

The purpose of this study is to develop a model of low-back mechanics and spinal loading which accounts for muscle coactivity, employing dynamically interpreted EMG and muscle kinematics. Limitations of previous models including inaccurate estimation of muscle coactivity as well as shear and torsion loading, misinterpretation of myoelectric activity and muscle capacity, artificial representations of lifting dynamics and input data have been addressed in this investigation. The aim is to predict accurately multidimensional trunk moments and spinal loads during iso-inertial lifting exertions, with valid and repeatable model performance and results.

BIODYNAMIC LIFTING MODEL

Model assumptions

The objective of our research is to attempt accurate modeling of the lumbro-sacral region, while avoiding as much unnecessary complexity as is practical within the constraints of validity. The model determines gross loading on the lumbar spine from dynamic muscle forces. Ten muscle equivalent vectors approximate trunk anatomy and mechanics.

Active muscle forces are assumed to describe adequately the trunk mechanics without consideration of passive muscle, ligament, and disc forces. The model is limited to trunk extension exertions within a range from 45° of flexion to the vertical. At extreme trunk flexion or extension, passive forces may become more significant, but this limited range of flexion may be modeled by active muscle forces (McGill and Norman, 1986). Gracovetsky and Farfan (1986) note that passive midline tension is small at these angles, but the thoracolumbar fascia may contribute extensor moment.

The muscle fibers sampled by the EMG electrode are assumed to be a reasonable approximation to the activity of the entire muscle body. EMG and muscle force are treated to be linearly related. The EMG to force relation has been debated in the scientific literature since the concept of quantitative electromyography was first popularized. Lippold (1952) and (later) Moritani and DeVries (1978) demonstrated linear relationships between voluntary isometric force and myoelectric signals detected with surface electrodes. Zuniga and Simons (1969) and Komi and Viitasalo (1976) proposed that surface EMG is proportional to the square of the isometric muscle force.

Recent studies have shown that processed EMG may be linearly related to force under appropriate strategies of muscle recruitment and rate coding (Solomonow *et al.*, 1990). Hof and Van Den Berg (1977) demonstrated that EMG is linearly related to muscle force, but coactivation generates a nonlinear relation between EMG and joint torque.

We use a linear relation between EMG and force for several reasons. First, the simplicity achieved by using a linear assumption makes the choice an expedient one. Second, over a small range of forces, a smooth, nonlinear relation may be validly approximated as linear. Third, the model accounts for muscle coactivation, thus, linear EMG to force relations are advocated based upon the work of Hof and Van Den Berg (1977). Finally, the muscle gain and EMG modulation algorithms employed in the model may be expanded in the future to account for nonlinear behavior should it prove necessary.

Model input

Required model input includes time-domain, dynamic data describing the kinetics, kinematics, and EMG levels of the exertion. Coefficients characterizing subject anthropometry, maximum EMG, and calibration values are employed to calibrate and format the dynamic data suitable for use in the model mechanics. A flow diagram of the model is presented in Fig. 1.

Trunk weight and center of mass are derived from Dempster's (1955) coefficients applied to measured anthropometry. Each muscle's cross-sectional area and lever arm vector are calculated as functions of trunk depth and breadth (Table 1) (Schultz *et al.*, 1982). Vector directions are assigned to the muscles based on the model of Schultz and Anderson (1981). Customized anthropometries may be employed by the model if MRI or CT data are available.

Externally applied, time-dependant forces are recorded from load cells located near each shoulder. Gravitational effects on trunk mass are calculated from trunk weight and flexion angle. The measured, external moments are used for comparison with the predicted, muscle-generated moments to determine

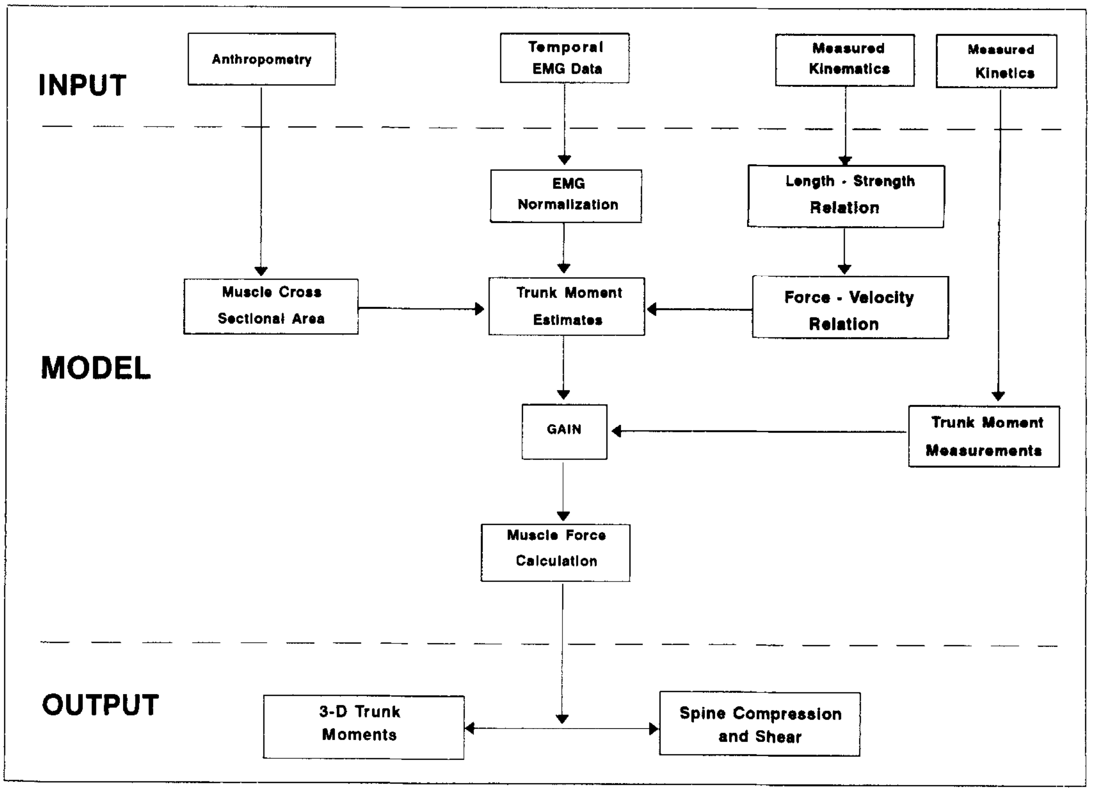

Fig. 1. Model flow chart.

Table 1. Coefficients for anthropometry

Muscle	Coefficients			Vector components		
	Area	Anterior moment arm	Rt lateral moment arm	X	Y	Z
Rt latissimus dorsi	0.0037	−0.28	0.21	−sin(45)	0	−cos(45)
Lt latissimus dorsi	0.0037	−0.28	−0.21	sin(45)	0	−cos(45)
Rt erector spinae	0.0389	−0.22	0.18	0	0	−1
Lt erector spinae	0.0389	−0.22	−0.18	0	0	−1
Rt rectus abdominis	0.0060	0.54	0.12	0	0	−1
Lt rectus abdominis	0.0060	0.54	−0.12	0	0	−1
Rt external oblique	0.0148	0.19	0.45	0	−sin(45)	−cos(45)
Lt external oblique	0.0148	0.19	−0.45	0	−sin(45)	−cos(45)
Rt internal oblique	0.0168	0.19	0.45	0	sin(45)	−cos(45)
Lt internal oblique	0.0168	0.19	−0.45	0	sin(45)	−cos(45)

Area = coefficient × depth × width
Anterior moment arm = coefficient × depth
Rt lat moment arm = coefficient × width

muscle strength parameters and for model performance characterization.

Dynamic measures of trunk bending angle and velocity are input into the model, and used to compute acceleration. The kinematic data are used to (1) describe the trunk motion as a function of time (2) calculate the acceleration loading effects on the trunk mass (3) determine the force and moment vector directions (4) modulate muscle EMG values to account for length and velocity artifact.

Kinetic and kinematic input data are graphically displayed as a function of time. The profiles are provided to verify whether the model had correctly accepted and interpreted the time-dependant data. Graphical output also provides the user with a representation of the modeled task.

Experimentally, the dynamic motion is restricted to controlled isokinetic or iso-inertial trunk extension in a vertical plane. The experimental apparatus (Fig. 2) physically constrains the vertical plane of motion and

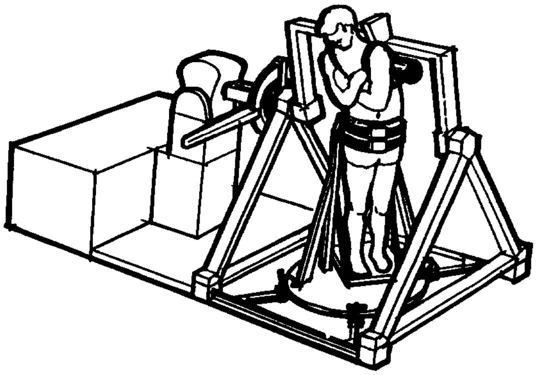

Fig. 2. Asymmetric reference frame (ARF).

may be asymmetrically rotated from sagittal about the vertical axis. The asymmetry is a constant for any given task, but may vary between tasks (Marras and Mirka, 1992a).

Ten channels of integrated myoelectric data are used by the model to calculate the relative muscle force. The time-domain, EMG values represent muscle activity from the left–right pairs of the Latissimus Dorsi, Erector Spinae, Rectus Abdominis, Internal Olique, and External Oblique. During the experiment, raw EMG signals were pre-amplified, high- and low-pass filtered at 80 and 1000 Hz, respectively, rectified, and integrated via a 20 ms sliding window hardware filter (Marras and Mirka, 1992a). Maximum and resting EMG values were collected as a function of flexion angle and asymmetry to normalize the dynamic EMG signals.

All dynamic data are smoothed via a Hanning weighted, time-domain filter. Smoothing the data is necessary to remove digitizing noise and artifact from differentiation and calibration routines. A Hanning filter was employed to exploit the sharp response characteristics demonstrated by this type of windowing while retaining processing simplicity.

The filter is assigned a 10 Hz equivalent noise bandwidth for processing kinetic, kinematic, and EMG data. The bandwidth was selected by comparing prefiltered and postfiltered EMG data, and agrees with the muscular physiologic tremor frequency (Lippold, 1970).

Muscle force

The tensile force generated by each muscle, j, is described by the normalized EMG, muscle cross-sectional area, a gain factor describing muscle force per unit area, and modulation factors describing EMG and force behavior as a function of muscle length and velocity.

$$\mathrm{Force}_j = \mathrm{Gain} \times \frac{\mathrm{EMG}_j(t)}{\mathrm{EMG}_{\mathrm{max}j}(\mathrm{Ang,\ Asmtry})} \times \mathrm{Area}_j \times F_j(\mathrm{Vel}) \times F_j(\mathrm{Ang}). \tag{1}$$

EMG data are normalized by myoelectric maxima from each muscle as a function of trunk bending angle and asymmetry. Normalization is necessary to remove interelectrode variability due to placement, skin abrasion, flesh resistance, muscle fiber density, and electronic channel differences. Normalizing as a function of both angle and asymmetry is necessary (Mirka, 1991) to avoid EMG error, and it is unique to this model. The data are expressed in units of percentage of the maximum.

$$\mathrm{NormEMG}_j = \frac{\mathrm{EMG}_j(t)}{\mathrm{EMG}_{\mathrm{max}j}(\mathrm{Ang,\ Asmtry})}. \tag{2}$$

$\mathrm{EMG}_j(t)$ is the time-dependent EMG signal level of muscle j, Ang represents the trunk flexion angle and Asmtry, the trunk asymmetry.

The EMGs are modified to account for the relation between tensile force and muscle length, $F_j(\mathrm{Ang})$. Motor unit density relative to muscle length, directly affects EMG voltage due to variation in the myoelectric potential density picked up by the surface electrode. The EMG signal may, therefore, contain artifact due to muscle length not related to muscle force. A unitless relation allowing modulation of the EMG data was developed by Marras and Sommerich (1991a) and is employed in the current model. The length–strength–modulation factor was based on data relating EMG level, muscle force, and trunk angle collected in our laboratory.

EMG values of each muscle are adjusted for velocity artifact, $F_j(\mathrm{Vel})$. Bigland and Lippold (1954) demonstrated that an increase in muscle contraction velocity yields increased myoelectric activity without a concomitant increase in force output. To avoid over-prediction of muscle force, EMG values are reduced by a ratio of the static to dynamic EMG levels, removing any velocity artifact in the signal. Both numerator and denominator of the velocity-modulation factors are a function of force and trunk flexion angle.

$$F_j(\mathrm{Vel}) = \frac{\mathrm{averageEMG}_j(\mathrm{Ang},\ F,\ \mathrm{Vel}=0)}{\mathrm{averageEMG}_j(\mathrm{Ang},\ F,\ \mathrm{Vel})}. \tag{3}$$

The coefficients for the modulation factor were developed from an extensive data base of EMG records collected at our laboratory, and approximate a theoretical form related to the Hill (1938) equation.

Normalized EMG values are multiplied by the respective muscle cross-sectional area, accounting for the relative force generating capacity of each muscle. It is presumed the muscle force capacity is directly related to cross-sectional area (Lamb, 1984). At this point, the processed EMG represents the percentage of the muscle cross-sectional area that is active at a particular point in time. The physiologic muscle-force per unit cross-sectional area, referred to as gain, may then be used to determine the dynamic tensile force of each muscle.

The muscle force per unit area, i.e. gain, is highly variable between subjects, based on subject conditioning, training, and natural ability. On the other hand, the gain for a given subject must be constant throughout each of the experimental trials. In this model, the subject's gain is computed by comparing the measured moment about the bending axis with the predicted moment generated by the muscle forces in a series of calibration tests. By Newton's second law, the internal extension moment must be equal to the sum of the measured, external moment and an acceleration term. Gain is appropriately and automatically adjusted to satisfy this condition.

For model-adequacy checking, the predicted gain level must fall within the physiological range 30–100 N cm^{-2} (McGill and Norman, 1987; Reid and Costigan, 1987; Weis-Fogh and Alexander, 1977). A subject's specific gain is calibrated based on several exertions averaged over a range of test conditions.

Mechanics

Axis directions are defined such that when the subject is standing in an upright, erect posture, the positive X-direction is right lateral, the Y-direction is anterior, and the Z-direction is cranial (Fig. 3). A coordinate reference frame is assigned to the sacral spine and another to the lumbar spine. The sacral reference system remains fixed in the laboratory frame. The lumbar frame moves with the trunk and describes the motion of the trunk relative to the sacral spine. System origins are co-located at the L5–S1 junction, and both systems are coincident only when the subject is standing erect.

A three-dimensional moment vector is calculated from the measured forces. The mass-acceleration term is grouped with the external moment vector for computational convenience. Vector components of external moment are resolved in the lumbar reference frame, then transformed (via Euler rotation) into the sacral coordinate frame for gain matching and analysis.

Since two load cells, one near each shoulder, were employed to collect force data, only two moment components may be computed. The third vector direction, i.e. moment about the Y-axis or lateral bending, is assigned a zero amplitude. As a result of vector rotation, lateral forces and moments may arise. Muscle forces vectors are computed from the normalized and modulated EMG data, muscle cross-sections, subject gain, and unit vector directions for all ten modeled muscle equivalents. A three-dimensional representation of the time-dependant, muscle activity at the L5–S1 level (Fig. 4) demonstrates the relative distribution of force among the trunk muscles. The schematic is useful for assuring proper input, normalization, modulation, and processing of the EMG data prior to mechanical analysis. Furthermore, the illustrated muscle activity allows (1) qualitative prediction of the internal moments generated by muscle tensile forces and (2) intuitive validation of mechanical results predicted by the model.

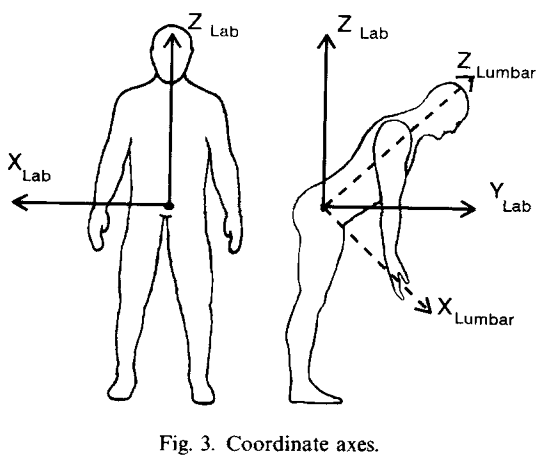

Fig. 3. Coordinate axes.

Muscle generated moments about the L5–S1 junction are predicted from vector products combining anthropometric moment arms with muscle forces. The moment calculation is

$$M_i = \left[\sum_{\text{muscle}} \{\mathbf{X}_j \times \mathbf{F}_j\} \right]_i , \tag{4}$$

where M_i is the moment vector component in the ith direction, X_j is the moment arm vector of the jth muscle and F_j, its force vector.

Predicted, internal, trunk moments are graphically displayed as a function of time superimposed upon the measured external trunk moments (Fig. 5). The shape of the predicted moment profiles must agree with the shape of the measured moment profiles if the model is correctly simulating trunk mechanics. Statistical comparison (by correlation) of the predicted and measured profiles provides a numerical estimate of the quality of the match between the two profiles.

Spinal loading, i.e. compression, right-lateral shear, and anterior shear forces, are calculated from the muscle equivalent force vectors. The relative distribution of force among the ten trunk muscles is determined from the modulated EMGs. If the predicted trunk moments agree with the measured values, then the predicted force amplitude in each of the muscles, and subsequent spinal loading must be correct. The compression and shear calculations are performed by summation of the vector components from each muscle,

$$L_i = \left[\sum_{\text{muscle}} \mathbf{F}_j \right]_i , \tag{5}$$

where L_i is the ith vector component of the load suffered by the spine at the L5–S1 junction and F_j, the tensile force vector in the jth muscle. Time-dependent force profiles of spinal compression, right-lateral shear, and anterior shear forces are graphically displayed (Fig. 6).

Measured and predicted values of the three trunk moments, predicted compression, anterior shear, and lateral shear are numerically displayed and written to a file for postmodeling analysis. The task or subject gain and statistical comparison of the measured and

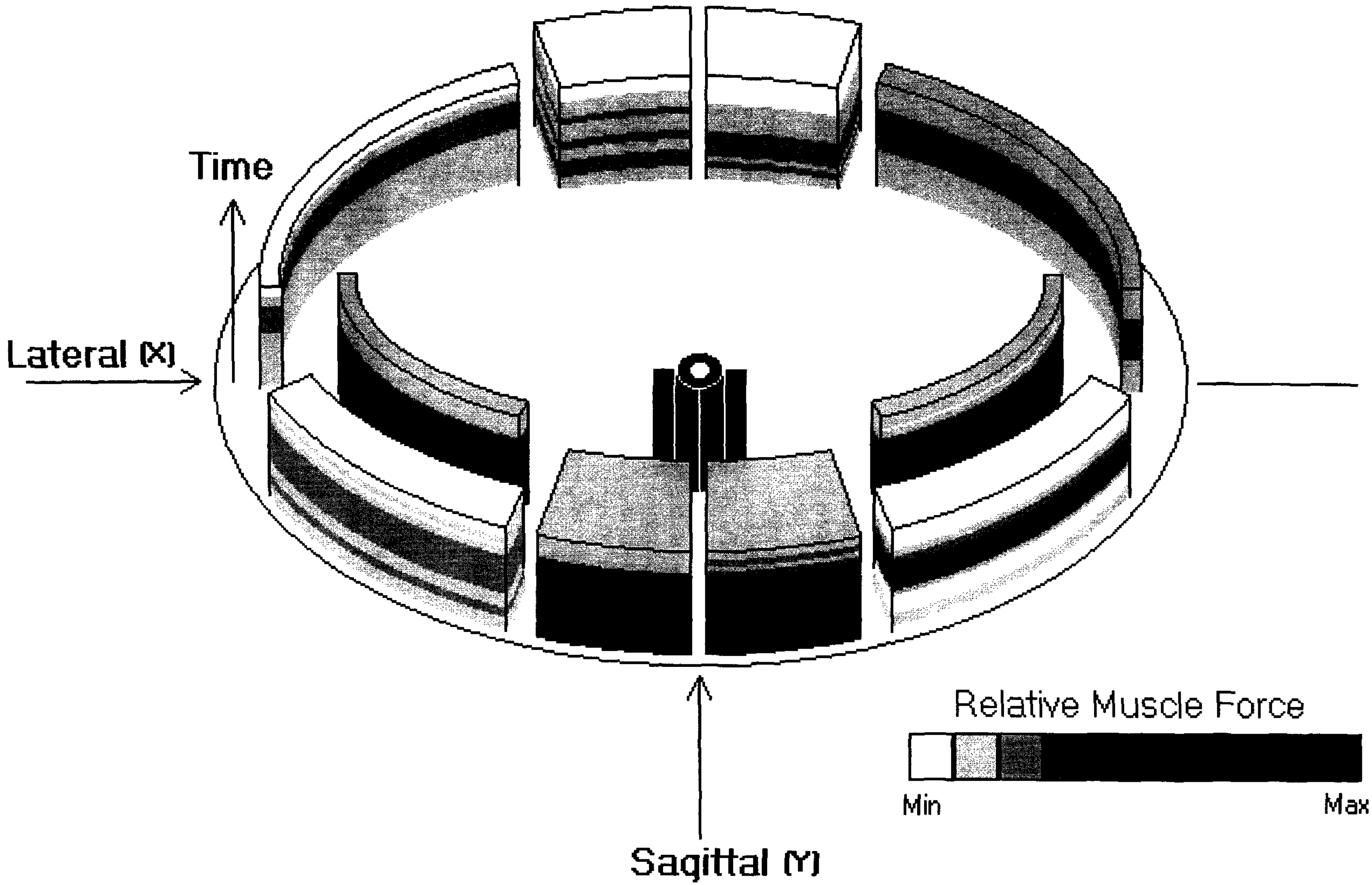

Fig. 4. Relative muscle force as a function of time (vertical axis) in each of the ten modeled trunk muscles viewed from the posterior aspect. Note the measured coactivity, especially in the antagonistic musculature. The illustration is a useful representation of muscle activity throughout a lifting exertion.

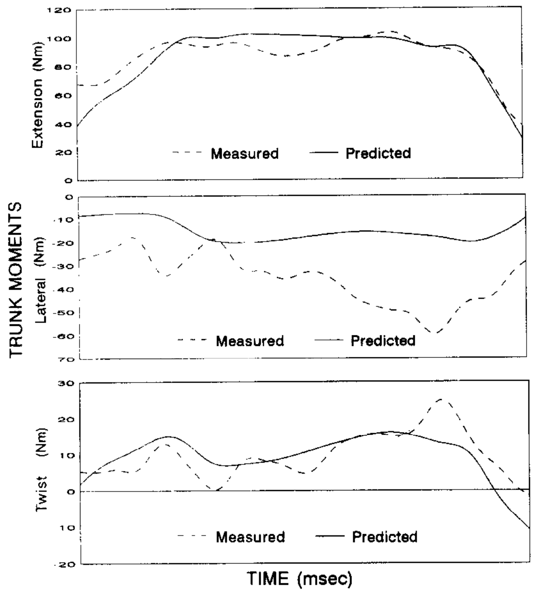

Fig. 5. Trunk moments measured as a function of time during a typical exertion (solid line) are superimposed on the trunk moments predicted from the EMG-assisted model (dashed lines). Note the modeled exertion moment profile closely predicts the measured profile, indicating the EMG-assisted model accurately predicts the dynamic behavior of trunk kinetics.

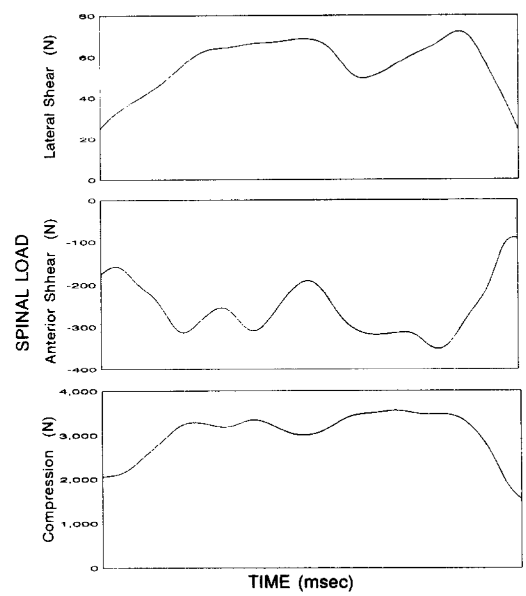

Fig. 6. Lateral shear force, anterior shear force and compression on the lumbar spine are predicted as a function of time from muscle activity determined via processed EMG data and calibrated subject gain. The illustration demonstrates the dynamic nature of the compression and shear loads on the spine throughout a lifting exertion.

predicted moment profiles are provided along with a kinetic task summary. The gain value may be used as a validity check so as to assure that physiologically reasonable muscle force per unit areas are predicted.

VALIDATION

The mean gain values calculated by the model and their distribution reveal that predicted results are physiologically reasonable. Modeled muscle force per unit area must fall within the physiologically acceptable range of 30 to 100 N cm^{-2}. Data from 20 subjects were processed and a distribution of subject gains were determined (Fig. 7). The average subject gain was 42 N cm^{-2} with a standard deviation of 11 N cm^{-2}. Thus, the muscle strengths predicted by the EMG-assisted model appear realistic.

Comparison of modeled and measured lifting moments illustrates the outstanding performance of the model. The square of the correlation coefficient is used to compare the lifting moment profiles on a point-by-point basis, and represents the variability accounted for in the model. An R^2 distribution has been achieved from a 2160 static, isokinetic, and isoinertial test trials under sagittally symmetric and asymmetric lifting conditions (Marras and Mirka, 1990), and this distribution demonstrates that over 80% of the trials performed with an R^2 greater than 0.8 (Fig. 8).

DISCUSSION

Previous biomechanical lifting models have employed physiologically unrealistic assumptions and generated inaccurate estimates of spine loading. The present model corrects the weaknesses of previous models by using conditioned EMG data to estimate dynamic trunk moments and three-dimensional spinal loading. The analysis employs an EMG-assisted model because it incorporates the neuromuscular control system of the trunk musculature through

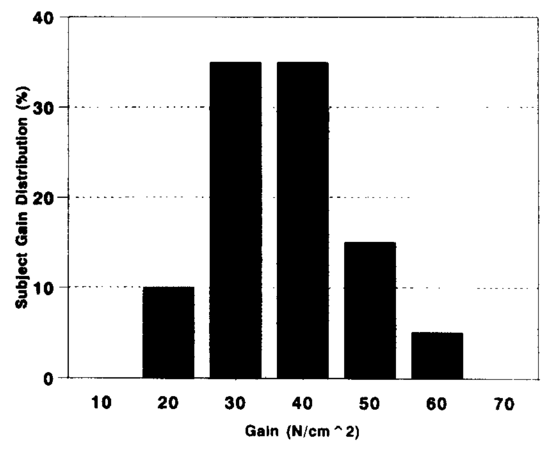

Fig. 7. A normalized distribution of calibrated gain (N cm^{-2}) from 20 subjects demonstrates the muscle force per unit area predicted by the model is physiologically reasonable. The muscle force per unit area was determined from test exertions performed by each subject. The gain was then used as a constant to quantify the muscle capacity in each of 108 exertions performed by the subject.

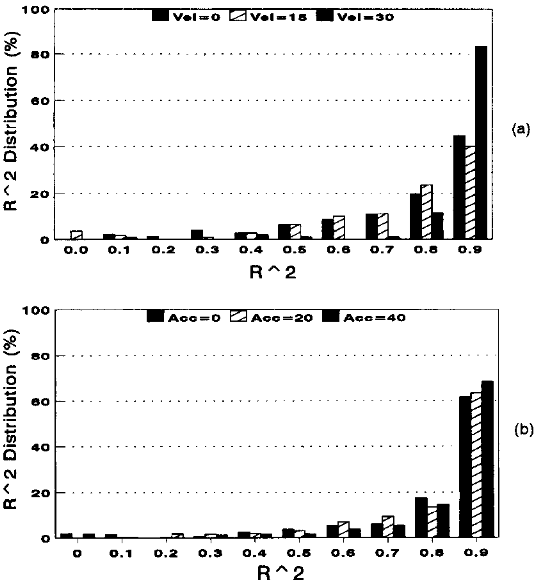

Fig. 8. Distribution of R^2 between predicted and measured performance for (a) isokinetic (0, 15, and 30° s^{-1}) and (b) isoinertial (0, 20, and 40° s^{-2}) lifting exertions. This distribution indicates that the model typically predicts over 80% of the variability in the 2160 trials.

direct measurement. Optimization models have attempted to predict muscular activity, but have yet to predict muscular antagonism successfully. Models which ignore muscle coactivity are incapable of correctly prédicting spinal compression and various components of shear and torsion loading. EMG-assisted models simply measure the myoelectric results of these complex control systems allowing accurate prediction of spinal loading.

This EMG-assisted model is an excellent predictor of trunk extension moment as illustrated by the subject gain levels (Fig. 7) and the R^2 distributions (Fig. 8). High average R^2 indicates that the model is capable of predicting the dynamic behavior of the extension moment accurately. The absolute magnitude of force in each muscle is derived from the measured trunk moments, while tensile force in each trunk muscle relative to others is determined by measurements of muscle activity (Fig. 4). The same muscle forces are used to predict trunk moments and spinal loads. Thus, if the model is capable of accurately predicting time-dependent, dynamic, extension moments (Fig. 5), it is assumed capable of accurately simulating spinal loading (Fig. 6).

As with any biomechanical analysis, the assumptions used to drive this model may be a limiting factor. One of the most significant assumptions effecting the model is that EMG is linearly related to muscle force. Note that the predicted moments approximate the measured levels independent of the moment amplitude (Fig. 5). This indicates that the linear EMG assumption is reasonable over the range of exertions examined.

The dynamic geometry of the spinal curvature has been reduced to a rudimentary flexion angle of the thorax relative to the sacrum. This assumption may affect the predicted trunk moments, the gain values achieved from those calculations, and the subsequent predictions of spinal loading. Recall that the model has been designed to represent accurately and realistically the mechanical loading and behavior of the low-back, while refraining from as much unnecessary complexity as possible. With these goals in mind, the model was not intended to predict forces on individual spinal vertebrae, but rather the gross loading in the lumbar region of the spine. Thus, a simple model of the spine was employed. Nevertheless, relative changes in spinal compression and shear as a function of trial kinematic and kinetic parameters remain valid. There is a need for future research to determine the extent of variability, if any, introduced into the model by the assumption of a straight and rigid spine.

Despite the challenges to the assumptions used to drive this model, the analysis demonstrates time-dependent, dynamic, trunk, extension moments may be predicted from kinematic and EMG data satisfying the constraints of physiological validity. Compression and shear forces at the lumbro-sacral junction may be predicted from the muscle equivalent forces generating the trunk moments. Therefore, qualitative (if not quantitative) loading may be compared as a function of lifting position, velocity, acceleration, and load. Trends predicted by the model agree with the dynamic relations proposed by Frievalds *et al.* (1984), McGill and Norman (1985), and Goel *et al.* (1991). Furthermore, the model has been tested and validated under conditions of controlled dynamics, generating results supporting epidemiological findings (U.S. Dept. Labor, 1982; Bigos *et al.*, 1986; Marras *et al.*, 1993).

The model developed in this study is capable of predicting trunk extension moments and spinal loads during asymmetric, dynamic, lifting exertions, and its results compare favorably with values directly measured about the lumbro-sacral region of the spine. Thus, it simulates dynamic loading of the spine under conditions of realistic lifting exertions without the need for muscle activity approximations. Its strengths are the insight it provide into (1) the effects of motion induced and (2) muscle co-activity on spinal loading. Consequently, the model produces a more realistic estimation of multidimensional, dynamic loading of the spine. Further benefit is derived from the ability of the model to predict dynamic shear loading as well as compression, for multidimensional spinal loading, i.e. shear force, in combination with compression which has been demonstrated to play a major role in the risk to low-back injury during lifting (Shirazi-Adl *et al.*, 1986; Shirazi-Adl, 1989).

CONCLUSIONS

An EMG-assisted, biodynamic, lifting model has been developed to simulate accurately lumbro-sacral extension moments generated by muscle forces during dynamic lifting tasks. Relative muscle force levels are determined from normalized EMG, modified to account for trunk position and muscle dynamics. Tensile magnitudes are computed from the relative force distribution among the trunk muscles and the dynamic extension moments. Spinal loading is predicted from the internal force vectors and external loading. A brief description of the model is provided as follows:

- The EMG-assisted model empirically determines bilateral muscle forces and coactivity.
- EMG data are normalized as a function of trunk flexion angle and asymmetry, then adjusted for subject-dependant, muscle length–strength and force–velocity factors.
- A fully dynamic representation of the lifting task is modeled and has been tested under controlled isometric, isokinetic, and iso-inertial exertions of sagittally symmetric and asymmetric conditions.
- Prediction of spinal loading is based on muscle tensile forces determined from EMG activity and scaled via solution of Newton's second law applied to trunk moment exertions.
- Muscle strength capacity, i.e. gain, is controlled as a subject-dependant constant, and prohibited from changing with each lifting task.

Acknowledgement—Funding for this study has been provided by the Ohio Bureau of Workers' Compensation, Division of Safety and Hygiene.

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