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AN EMG-ASSISTED MODEL OF TRUNK LOADING DURING FREE-DYNAMIC LIFTING

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Abstract—One of the continuing challenges in biomechanics has been to assess loading of the spine during dynamic lifting exertions. A model was developed to accurately simulate multi-dimensional spinal loads and trunk moments from measured muscle coactivity and external forces during free-dynamic lifting exertions. Model validity was demonstrated by comparing measured and predicted trunk extension moments. Its purpose was to examine realistic representations of lifting kinetics, kinematics, and dynamic trunk mechanics that may influence spinal loading, and to demonstrate that EMG-assisted modeling techniques can be applied to the analysis of free-dynamic exertions.

Spinal loads and trunk moments were predicted from the muscle force vectors and external loads. Muscle tensile forces were determined from the product of normalized EMG data modulated to account for contractile dynamics, muscle cross sectional area, and muscle force per unit cross-sectional area. Model output was physiologically valid, i.e. average predicted muscle force per unit cross-sectional area of 50–65 $\mathrm{N\,cm^{-2}}$, and accurately predicted measured, dynamic, lifting moments, with an average $R^2 = 0.81$ in the sagittal plane and $R^2 = 0.76$ in the lateral plane. Results indicated that compressive and shear loading increased significantly with exertion load, lifting velocity, and trunk asymmetry.

Keywords: Spine; EMG; Muscle; Trunk; Lifting.

INTRODUCTION

Accurate analyses of biomechanical loads on the lumbar spine during lifting exertions require a model capable of realistically representing both trunk muscle activity and lifting exertion dynamics. Spinal loads that occur during lifting tasks are significantly influenced by lifting dynamics (Freivalds *et al.*, 1969; Goel *et al.*, 1991; McGill and Norman, 1985; Marras and Sommerich, 1991b) and muscle coactivity (Hof and Van Den Berg, 1977; Marras and Mirka, 1992; Zetterberg *et al.*, 1987). Optimization based, biomechanical models of lifting have been developed that attempt to simulate realistic lifting dynamics (Freivalds *et al.*, 1984; Goel *et al.*, 1991; Kromodihardjo and Mital, 1986), but fail to represent muscle coactivity accurately (Hardt, 1978; Marras, 1988), thereby underestimating spinal load (Granata and Marras, 1994). Validated models employing electromyographic measures, i.e. EMG-assisted models, are capable of accurately representing muscle coactivity, but until now have been limited to static, isokinetic, or restricted motion representations of lifting (Granata and Marras, 1993; McGill and Norman, 1986; Marras and Sommerich, 1991a).

Many existing biomechanical models of trunk loading are compelled to neglect significant muscle coactivity due to mathematical constraints. Deterministic models have attempted to predict muscle activity via objective functions designed to minimize spinal compression (Kromodihardjo and Mital, 1986; Schultz and Andersson, 1981), spinal stress (Gracovetsky and Farfan, 1986; Schultz *et al.*, 1987), muscle stress (Goel *et al.*, 1991; Laddin *et al.*, 1989; Schultz *et al.*, 1982a), and combinations of these (Bean *et al.*, 1988; Schultz *et al.*, 1982b). However, the nature of the modeled coactivity is often not consistent with observed levels of muscle activity during lifting exertions (Marras and Mirka, 1992; Zetterberg *et al.*, 1987), and they typically neglect antagonistic muscle activity.

EMG-assisted models represent muscle coactivity and variability via direct measurement but have been limited in their ability to interpret force from myoelectric activity of dynamically contracting muscles. Dynamic trunk loads have been derived by modulating the myoelectric input data via theoretical (McGill and Norman, 1986; McGill, 1992) and empirical (Granata and Marras, 1993) length–strength and force–velocity relations. However, successfully relating myoelectric activity to time-dependent muscle force and joint torque during free-dynamic lifting has not been validated in the reviewed literature. In fact, Sommerich and Marras (1992), discourage correlating EMG amplitude with muscle tension unless kinematic controls are provided. Conversely, from Sudhakar (1990), Yang *et al.* (1991) and Marras and Mirka (1992), one may conclude that relative muscle force may be determined from EMG activity, if inertial dynamics and muscle coactivity are considered. Therefore, it may be possible to simulate free-dynamic trunk and spinal loads from EMG-assisted modeling techniques.

The purpose of this study was to demonstrate that measured EMG activity may be employed to represent muscle forces during free-dynamic exertions, and to develop a validated model of low back mechanics to simulate spinal loads experienced during smooth, free-dynamic, lifts. The objective was to accurately predict

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multi-dimensional, dynamic, trunk moments and spinal loads, with valid and repeatable model performance, while avoiding as much unnecessary complexity as possible.

METHODS

The model employs EMG and kinematic input to determine the dynamic, relative muscle force vectors of the ten, modeled, trunk muscles including the right and left latissimus dorsi, erector spinae, rectus abdominis, internal abdominal obliques, and external abdominal obliques (Schultz and Andersson, 1981). The tensile force generated by each muscle, j (equation (1)), is determined from 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 behavior as a function of muscle length $f(\text{length}_j)$, and velocity, $f(\text{Vel}_j)$ (McGill and Norman, 1986; Marras and Sommerich, 1991a):

$$\text{Force}_j = \text{Gain}\frac{\text{EMG}_j(t)}{\text{EMG}_{max\,j}}\text{Area}_j\, f(\text{Vel}_j)\, f(\text{Length}_j). \quad (1)$$

EMG data are normalized relative to myoelectric maxima collected during maximum voluntary contraction (MVC) efforts. This was necessary to remove possible analytical errors related to electrode placement, skin abrasion, flesh resistance, muscle fiber density and depth, and electronic channel differences (Mirka, 1991).

Relative myoelectric activities are multiplied by unitless functions of length, $f(\text{length})$, and velocity, $f(\text{Vel})$, to incorporate physiologic, force–length and force–velocity relations into the model (Bigland and Lippold, 1954; Vredenbregt and Rau, 1973). The functional coefficients were determined from preliminary calibration trials by minimizing the average variation of gain predicted by the model as a function of length and velocity. The modulation factors (equations (2) and (3)) employ the instantaneous length and velocity of each muscle, j, determined from anthropometric coefficients and kinematic inputs and normalized to their resting lengths.

$$f(\text{Length}_j) = -3.2 + 10.2\,\text{Length}_j - 10.4\,\text{Length}_j^2 + 4.6\,\text{Length}_j^3, \quad (2)$$

$$f(\text{Vel}_j) = 1.2 - 0.99\,\text{Vel}_j + 0.72\,\text{Vel}_j^2. \quad (3)$$

Empirically determined coefficients for the length-modulation factor agree with a Taylor expansion of the theoretical relation proposed by McGill and Norman (1986) and McGill (1992). The velocity-modulation factor is constrained to a maximum value of 1.2 for eccentrically contracting muscles and approximates the relationship represented by the Hill (1938) equation.

The physiological cross-sectional area, origin and insertion of each muscle is calculated as a function of subject anthropometry, including trunk depth and breadth (McGill *et al.*, 1988, 1993; Schultz *et al.*, 1982a) and dynamically updated via measured trunk posture and motion. Normalized EMG data are multiplied by their respective muscle cross-sectional areas to account for the relative force generating capacity of each muscle (Close, 1972; Lamb, 1984). The three-dimensional vector direction of each muscle is determined from its instantaneous endpoint positions. Force vectors are represented by the tensile magnitude, i.e. equation (1), and the vector direction of each muscle equivalent.

Gain, i.e. muscle force per unit area, is computed by comparing muscle-generated, trunk moments with measured, applied moments about the lumbo-sacral junction. Gain is appropriately and automatically adjusted to satisfy the equations of dynamic equilibrium. Thus, the absolute magnitude of the modeled force in each muscle is derived from the measured trunk moments, while relative tensile force in each trunk muscle is determined from measurements of muscle activity. To be physiologically valid, predicted gain values must lie between 30 and 100 N cm^{-2} (McGill and Norman, 1987; Reid and Costigan, 1987; Weis-Fogh and Alexander, 1977). Muscle force per unit urea 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 provides a means for testing model validity.

Trunk moments are computed from dynamic, muscle force vectors and anthropometric moment arms. Moments predicted by the EMG-assisted model are displayed as a function of time and superimposed upon the trunk moments measured from the force plate (Fig. 1).

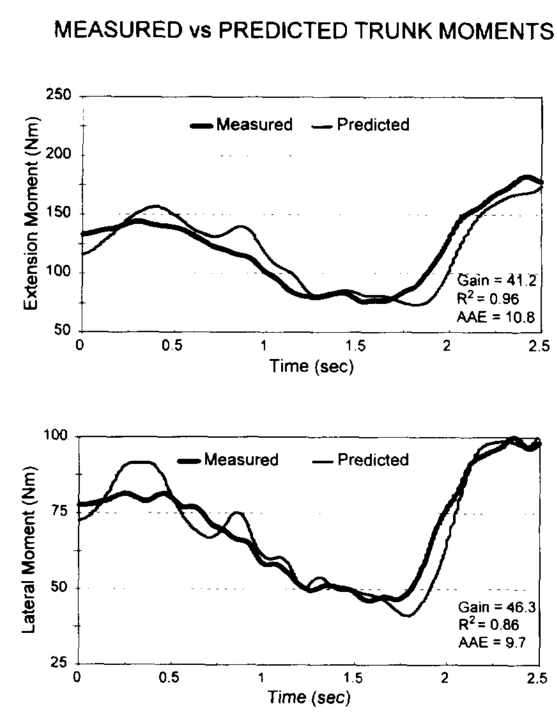

Fig. 1. Validation of the dynamic response of modeled output was performed by qualitatively and statistically (R^2) comparing measured (thick line) and predicted (thin line) trunk extension moments of each lifting trial. Note the modeled moment closely simulated the measured profile.

Model accuracy is examined by comparing the predicted and measured moment profiles and quantitatively determined by means of a statistical correlation value (R^2) and average absolute error (AAE). A high R^2 value, together with a valid subject gain value, indicates that the free-dynamic model accounts for the variability in the lifting moment, and implies that the model generates an accurate simulation of spinal load during the lifting exertions. The AAE value indicates the magnitude of the error between the simulated and measured trunk moments.

Spinal load, i.e compression, right-lateral shear, and anterior–posterior shear forces, are calculated by vector summation of the muscle equivalent force vectors, trunk weight and external loads on the trunk (Fig. 2). Measured and predicted values of the trunk moments, as well as predicted compression, posterior shear, and lateral shear forces are written to a file for post-modeling analysis. The task gain and correlation between measured and predicted moment profiles are recorded for model performance evaluation.

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 healthy males 25–31 yr old, with mean height of 177.5 ± 13.4 cm and mean weight of 76.8 ± 8.3 kg voluntarily participated in this study. Analysis indicated nine subjects were necessary for a statistical power of 0.80 at the $\alpha = 0.01$ level. None of the subjects had any history of low-back disorders, and all

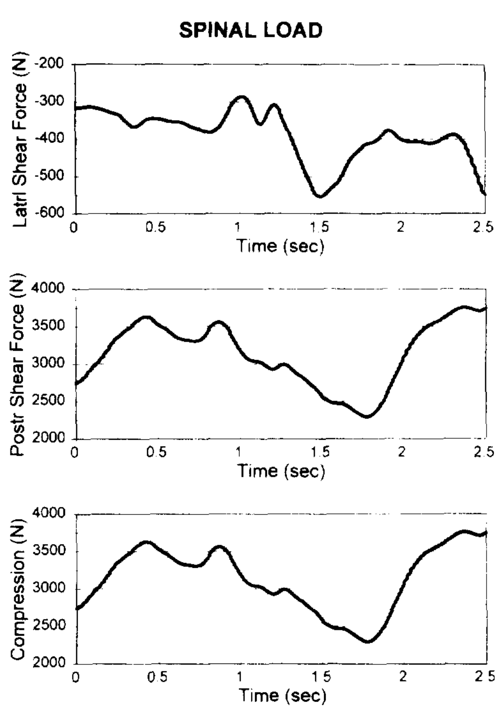

Fig. 2. Time dependent lateral shear force, posterior shear force, and compression at the lumbo–sacral junction were modeled from muscle activity determined via processed EMG and calibrated subject gain.

subjects were informed about the nature of the experiment prior to the test.

Subjects lifted loads of 18.2, and 36.4 kg at isokinetic trunk angular velocities (0, 30, 60 and 90 deg s^{-1}) as well as free dynamically (slow, medium and fast lift rates). While standing in an upright posture, the legs and pelvis of each subject were fastened to a rigid structure extending from the force plate. Isokinetic lifts were performed from a flexed trunk position of 45° to an erect posture at a rate subjectively controlled from video feedback. To perform free-dynamic exertions subjects were instructed to complete the entire motion in 2 s (free-dynamic slow), 1 s (free-dynamic medium), or as quickly as possible without jerking (free-dynamic fast). Lift weights and isokinetic velocities were selected to compare modeled results to prior studies (Granata and Marras, 1993; Marras and Mirka, 1992) and to represent lifting conditions typical of low and high risk manual materials handling industries (Marras *et al.*, 1993).

Voluntarily applied external kinetics, including gravitational moments and acceleration effects on trunk mass were dynamically measured by a force plate. Translation of force plate mechanics was performed to compute three-dimensional force and moments about the known location of the lumbo-sacral spine. Trunk position, velocity and acceleration were computed from dynamic measures of trunk flexion, twist, and lateral angles collected from a lumbar motion monitor (LMM) (Marras *et al.*, 1992). Integrated, myoelectric data were collected from the right and left latissimus dorsi, erector spinae, rectus abdominis, internal abdominal obliques, and external abdominal obliques (Mirka and Marras, 1993). EMG signals were collected from surface electrodes, 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, 1990). 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 filter was assigned a 10 Hz equivalent-noise-bandwidth for processing all data. The bandwidth was selected by comparing pre-filtered and post-filtered EMG data, assuring the filter removed high frequency noise while retaining signal dynamics. The 10 Hz bandwidth agrees with the physiologic muscular tremor frequency measured by Lippold (1970).

RESULTS

Subject gain values averaged over all dynamic exertions were 64.9 ± 27.6 N cm^{-2} in the sagittal plane and 50.2 ± 31.9 N cm^{-2} in the lateral plane. These values fall within the physiologically acceptable range of 30–100 N cm^{-2}. Because a subject's muscle strength per unit area cannot change from one exertion to the next, a subject's gain value must remain constant. Although gain changed significantly between subjects, the values did not vary significantly within subjects ($p < 0.01$). Gain values were not affected by lifting velocity or exertion level, although it was influenced by task asymmetry (Table 1). Thus, the model predicted muscle forces per unit cross-sectional area which were physiologically-valid, subject-dependent constants independent of the lifting force and velocity.

Distributions of squared correlation coefficients were achieved from the dynamic, lifting trials (Fig. 3). Sagittal plane lifting moments were predicted with an average R^2_{sagtl} of 0.81 and lateral moments with an average R^2_{latrl} of 0.76. Statistical analysis of variance (ANOVA) demonstrates that the EMG-assisted model accurately simulated trunk moment dynamics independent of the lifting exertion (Table 1). The analyses of squared correlation coefficients illustrate the model performed well at all dynamic velocities. Average absolute error (AAE) was employed to represent the average magnitude difference between measured and predicted dynamic moments. There was no statistical difference between the isokinetic (average $R^2_{\text{sagtl}} = 0.81$, $R^2_{\text{latrl}} = 0.74$) and free-dynamic exertions (average $R^2_{\text{sagtl}} = 0.82$, $R^2_{\text{latrl}} = 0.78$). Correlations between measured and predicted lifting moments were significantly ($p < 0.01$) reduced ($R^2_{\text{sagtl}} = 0.24$, $R^2_{\text{latrl}} = 0.26$) during isometric exertions. Low isometric R^2 values resulted from an unexplained cyclic variation in the extension moment predicted by the EMG-assisted model. The amplitude of the predicted variability was small, characterized by an isometric AAE of 5.8 N as compared with an average dynamic AAE of 21.5 Nm. The model simulated sagittal trunk moments to within an average absolute magnitude of 17.5 Nm; an error less than 15% of the peak lifting moment. Lateral trunk moments were stimulated with an AAE of 14.6 Nm, representing 24% of the peak lateral moments. Thus, the model predicted physiologically valid and kinetically accurate results for all dynamic, lifting exertions.

Although the values of the lifted weights were held constant across conditions, the average applied trunk moment decreased significantly with lifting velocity (Fig. 4). To avoid confounding the velocity and exertion level parameters, spinal loads were normalized by the applied trunk moment prior to analysis. Relative spinal loads generated throughout dynamic, lifting exertions increased as a function of trunk asymmetry (Fig. 5) and lifting velocity (Fig. 6). ANOVA of model output (Table 2) demonstrates that the lifting moment, lateral shear force, posterior shear force, and spinal compression were significantly ($p < 0.01$) influenced by exertion load (FORCE), lifting velocity and task asymmetry.

DISCUSSION

Lifting tasks generate trunk and spinal loads that are significantly influenced by lifting dynamics and muscle coactivity (Freivalds *et al.*, 1984; Granata and Marras, 1994). Analyses that attempt to model the biomechanical loads on the trunk and spine must, therefore, simulate realistic lifting exertions and accurately represent muscle co-contraction.

The EMG-assisted model developed in this study successfully predicted free-dynamic, trunk moments as illustrated by physiologically valid and stable subject gain levels, high average R^2 (Fig. 3) and low average absolute error values. High average squared correlation coefficients indicate that the model is capable of accurately predicting the dynamic behavior of extension moments under isokinetic and free-dynamic lifting conditions. Low AAE values indicate that the magnitude of the predicted variability is less than 15% of the task maximum moment in the sagittal plane; this is higher for lateral moment simulations.

Although the lifting model simulates trunk moments and spinal loads in three dimensions, only extension moments were validated for sagittally symmetric exertions. Applied twisting and lateral moments were minimal

Table 1. Statistical ANOVA of performance parameters indicate the model performed well under a wide variety of lifting tasks. Muscle force per unit area (Gain) and the correction between measured and predicted extension moment (R^2) during dynamic lifting exertions indicate the model validity and accuracy were not significantly influenced by lifting velocity or exertion level at $p < 0.01$. There were no significant differences in model performance during isokinetic and free-dynamic lifting exertions

	$\text{Gain}_{\text{Sagtl}}$	R^2_{Sagtl}	$\text{Gain}_{\text{Latrl}}$	R^2_{Latrl}
ASMTRY	$p <$ [illegible]	$p < 0.750$	—	—
Velocity ($V > 0$)	$p < 0.847$	$p < 0.849$	$p < 0.293$	$p < 0.298$
Force ($F > 0$)	$p < 0.543$	$p < 0.345$	$p < 0.058$	**$p < 0.014$**
Isokin vs. free-dyn	$p < 0.319$	$p < 0.873$	$p < 0.250$	$p < 0.071$

ANOVA effects significant at $p < 0.01$ have been shaded.
ANOVA effects significant at $p < 0.05$ have been noted in bold.
NB, no significant interactions.

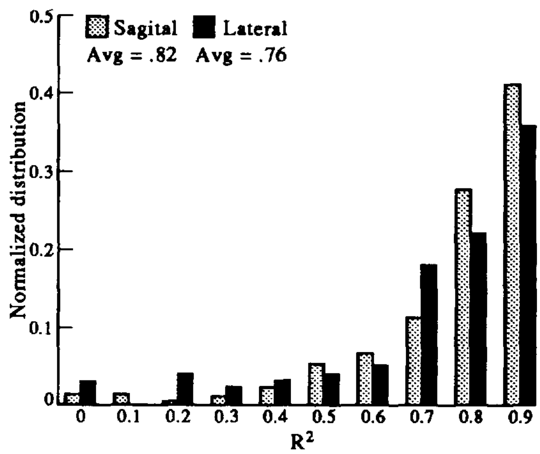

Fig. 3. A distribution of the squared correlations between measured and predicted lifting moments indicates the model accurately predicted trunk extension dynamics during sagittally symmetric and asymmetric exertions. R^2 values representing lateral plane response were achieved during asymmetric lifting exertions. No significant differences between isokinetic and free-dynamic exertion results were found under the test conditions.

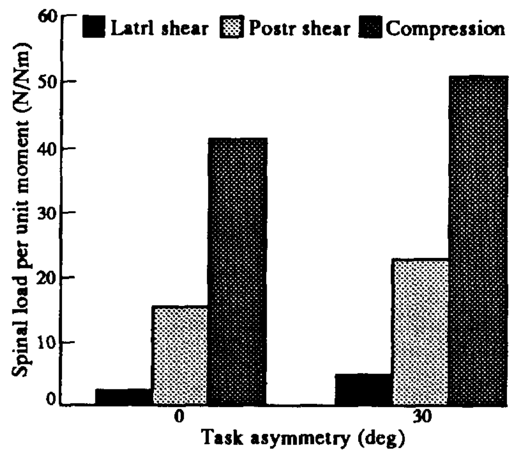

Fig. 5. Spinal load per unit extension moment is plotted as a function of lifting task asymmetry. The relative lateral shear force (F_x/M_x), relative posterior shear force (F_y/M_x), and relative compression (F_z/M_x), increased with asymmetry. Increased spinal loads were statistically significant in all three dimensions.

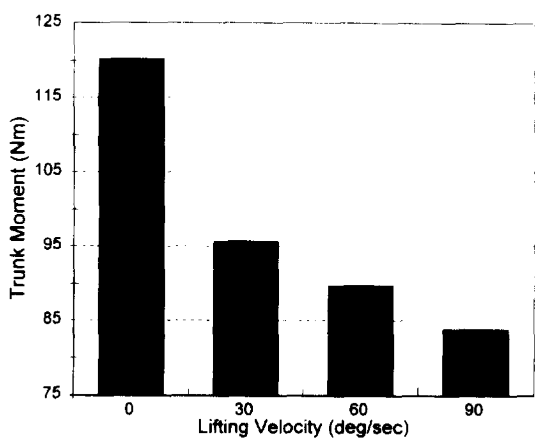

Fig. 4. Despite lifting loads of constant weight, applied trunk moments decreased significantly with lifting velocity. As subjects were required to increase the velocity of the lift, they appeared to pull the weight close to their center of balance earlier in exertion than during slower lifts.

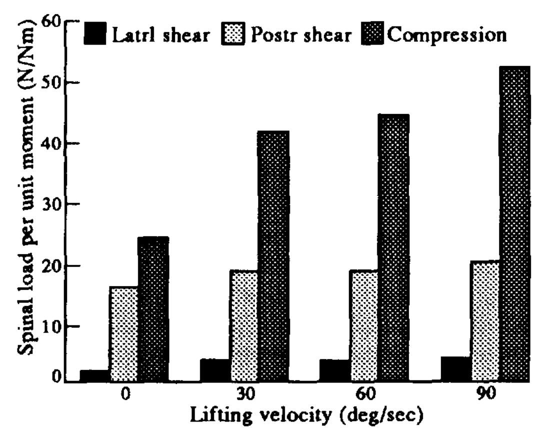

Fig. 6. Spinal load per unit extension moment is plotted as a function of lifting task velocity. The relative lateral shear force (F_x/M_x), relative posterior shear force (F_y/M_x) , and relative compression (F_z/M_x), increased with extension velocity. Increased relative lateral and posterior shear loading were statistically insignificant.

during those tasks introducing signal to noise errors. During maximal extension exertions, average coupled lateral moments equal to 10% of the extension moment and average twisting moments equal to 7% of the extension moment have been reported by Parnianpour *et al.* (1991). Small, e.g. 1–2 Nm, perturbations in the predicted lateral and twisting moment may amount to as much as 40% of the measured coupled moments during extension exertions.

To assess trunk loading during asymmetric lifting exertions, the biomechanical analyses must perform well in both the sagittal and lateral planes. Results demonstrate that during asymmetric exertions, the model performed well in simulating the lateral as well as sagittal trunk moments. Applied twisting moments were still too small during asymmetric lifts to model this accurately. A recent study by Thelen *et al.* (1994) succeeded in implementing an EMG-assisted model which predicted isometric trunk moments in multiple simultaneous dimensions. The free-dynamic EMG-assisted model is the first to realistically simulate multi-dimensional trunk moments generated during dynamic exertions via measured coactive muscle behavior. Future research shall focus on applying and measuring significant lateral, twisting and coupled moments to evaluate the model performance in three simultaneous dimensions.

Validation of the biomechanical analyses was performed by examining the predicted muscle force per unit area, i.e. gain, the average absolute error, AAE, and

Table 2. Statistical ANOVA demonstrate spinal compression per unit trunk moment (F_Z/M), lateral shear force per unit moment (F_X/M) and posterior shear forces per unit moment (F_Y/M) were significantly influenced by task asymmetry. Lifting velocity and load influenced relative spinal compression and applied lifting moment (M)

	M	F_X	F_Y	F_Z
ASMTRY	$p < 0.336$	[illegible]	[illegible]	[illegible]
Velocity	[illegible]	$\mathbf{p < 0.016}$	$\mathbf{p < 0.021}$	[illegible]
Force	[illegible]	$\mathbf{p < 0.083}$	[illegible]	[illegible]
Asmtry × vel	$p < 0.653$	[illegible]	$p < 0.853$	$p < 0.353$

ANOVA effects significant at $p < 0.01$ have been shaded.

ANOVA effects significant at $p < 0.05$ have been noted in bold.

NB, only significant interactions noted.

correlation and between measured and predicted trunk moments, R^2. These three parameters must be employed as a group to judge the performance and validity of a lifting model. Gain represents the muscle stress required to generate the magnitude of the average lifting moment, without reference to the model's ability to simulate the shape or variability of the predicted kinetics. Conversely, correlation coefficients may indicate excellent agreement between measured and simulated dynamics, without reference to the relation between the magnitudes of the moments represented by the simulation. Average absolute error provides a measure of error magnitude not inferable from the gain and R^2 statistics. Thus, all three performance measures must be examined to assure model validity.

The performance of the free-dynamic model was validated throughout the entire range of motion of a given lifting task. Measured and predicted moment profiles were compared on a point by point (100 Hz) basis over a time span as long as 5 s. Previous EMG-assisted models (McGill and Norman, 1986; Reilly and Marras, 1989) attempted to simulate trunk loading during constrained static and isokinetic exertions. Validation of those models was not presented, i.e. measured trunk moments were not compared with predicted values. Models by Marras and Sommerich (1991a, b) and Granata and Marras (1993) permitted model validation by direct comparison of predicted and measured extension moment values, but represented constrained isokinetic and isoinertial exertions validated over a small range of angles, $\pm 2.5°$. Comparisons of measured and predicted kinetics throughout the entire range of motion were not reported. Furthermore, those analyses represented constant extension moments as opposed to lifting exertions. Results from the free-dynamic model demonstrate that it is possible to construct and validate an EMG-assisted model which reasonably simulates trunk kinetics throughout a broad range of trunk angles during free-dynamic lifting exertions.

Spinal loads predicted by the model cannot be validated by measurement, but the same mathematical set is employed to simulate trunk moments and spinal loads, i.e. the same muscle forces used to predict trunk moments are used to predict spinal loads. It is clear that an inability to reasonably simulate trunk moments, precludes accurate prediction of spinal loads. Neglect of muscle coactivity similarly generate unrealistic representations of spinal loads (Granata and Marras, 1994). Although, accurate representations of coactive trunk moments does not ensure accurate spinal load predictions, it demonstrates improved model validity and increased confidence in the values of modeled spinal loads. Therefore, since the model is capable of accurately predicting free-dynamic lifting moments via measured muscle activity (Fig. 1), we assume that it accurately represents the influence of realistic lifting parameters upon relative spinal loads (Fig. 2).

Model results demonstrated that spinal load increase with the velocity and asymmetry of the task as well as the weight of the load as cited in previous analyses of dynamic exertions. Studies by Freivalds *et al.* (1984), McGill and Norman (1985), Goel *et al.* (1991), and Marras and Sommerich (1991b) conclude that spinal compression increases with trunk extension velocity. These results support epidemiological findings that document risk of low-back pain is related to exertion load, velocity and asymmetry (U.S. Dept. Labor, 1982; Bigos *et al.*, 1986; Marras *et al.*, 1993).

Since the applied trunk moment changes with lifting velocity (Fig. 4), comparing spinal load from different velocities is confounded by the fact that the exertion level also influences spinal load. Therefore, the spinal load was normalized by the magnitude of the vector sum of trunk moment, e.g. compression per unit of applied trunk moment. Despite the fact that exertion variability was removed, lifting velocity significantly influenced relative spinal compression and anterior–posterior shear force (Fig. 6). Task asymmetry affected the relative spinal loads in all three dimensions (Fig. 5).

Research indicates that spinal load increases with velocity and task asymmetry due to muscle coactivity. Studies (Marras and Mirka, 1992) have shown that muscle coactivity increases with lifting velocity and trunk asymmetry. Coactivity, especially antagonistic coactivity, can significantly increase the biomechanical load on the spine without increasing the applied trunk moment (Granata and Marras, 1994; Hughes, 1991). Considering the fact that relative spinal load increased with trunk kinematics despite being normalized for exertion level supports the belief that increased coactivity necessary for dynamic trunk stabilization increases spinal load.

The distributions of correlation coefficients (Fig. 3) demonstrate that processed, and carefully modulated EMG can represent dynamic muscle force over a wide range of lifting velocities. Furthermore, there was no statistical difference between isokinetic and free-dynamic lifting exertions; both performed equally well. Model input, specifically muscle activity patterns represented by EMG data, appear to be similar enough between isokinetic and free-dynamic lifting exertions in which model performance was not affected. Controlled but unconstrained lifting exertions may, therefore, be employed to represent accurately the biomechanics of free-dynamic lifting under these conditions.

Appropriate representation of muscle area is essential to the validity and performance of the biomechanical model. Relative magnitudes of muscle contractile forces were computed from processed EMG and scaled by the physiological cross-sectional areas. Previous EMG-assisted models (Granata and Marras, 1993; Marras and Sommerich, 1991a; Reilly and Marras, 1989) employed muscle areas representative of the cross-sections found at the transverse plane through the lumbo–sacral junction (Schultz and Andersson, 1981). We have determined that the latissimus dorsi is a large, powerful muscle which is poorly represented by the area of a few slips of muscle fiber that pass through the lower lumbar levels. Consequently, the maximum cross-sectional area of the latissimus dorsi, found near T5 (McGill *et al.*, 1993), was found to be more appropriate for modeling the force generating capacity of that muscle. Similar determination of muscle area were used to describe the other muscles included in the model.

Despite the fact that muscle lines of action change with trunk position, previous biomechanical models of lifting (Granata and Marras, 1993, Reilly and Marras, 1989, Schultz and Andersson, 1981) assumed that muscle vector directions were constant in space. The free-dynamic model was designed to permit each muscle orientation, length, and velocity to move with the lifting motion and position of the trunk. This was accomplished by dynamically locating muscle origins and insertions via Euler rotation of 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 a typical exertion.

Posture-dependent muscle lengths and orientations more realistically represent biomechanical dynamics but were limited by the assumption of straight line vectors between endpoints. The mechanical structure of the model has been developed by rotating muscle insertion positions relative to their origins from measured trunk motion data. The goniometer (LMM) used to monitor trunk kinematics measures the relative position of the thorax relative to the pelvis (Marras *et al.*, 1992). This method bypasses the need to determine point translation through the complex geometry of the spine. Conversely, the three-dimensional nature of the trunk and its ability to maintain trunk angle with and without spinal lordosis complicate the computation of muscle length and associated velocity. Furthermore, describing the vector direction from the location of the muscle end-points overlooks curvature lengthening. The limitations describing muscle length may indicate the source of significant variation of the sagittal gain values described in Table 1. Although the inclusion of dynamically varying muscle kinematics and vector directions significantly improves the realism of modeled dynamics, improved geometric modeling of the muscles may further enhance the accuracy of dynamic models.

Throughout this research, it has been assumed that 10 muscle vectors adequately describe trunk anatomy and mechanics. Discriminate function analyses indicated that subsets of the 10 muscles employed in this study were capable of predicting coactive trunk moments for a given task. All 10 of the muscle equivalents were necessary to simulate lifting moments accurately under a variety of lifting conditions. Research has shown (Granata and Marras, 1994) that neglecting any of these coactive muscles can significantly influence the predicted value of relative spinal load. Although results indicate that the 10 muscles employed in this study can be employed to simulate trunk extension moments accurately, one might be tempted to include more muscle equivalents to achieve a slight increase in kinetic accuracy. The power of an EMG-assisted model lies in its ability to empirically determine muscle coactivity during an exertion. Therefore, inclusion of muscles, whose myoelectric measurement cannot be achieved, reduces the modeled significance of coactive results, thereby reducing the accuracy of modeled loads.

Muscle forces were assumed to be an adequate description of trunk mechanics without consideration of ligament and disc restorative moments. At extreme flexion–extension angles, passive forces may become significant (Gracovetsky and Farfan, 1986), but within the design range of 45° flexion to vertical, the trunk moments may be represented by active muscle forces (McGill and Norman, 1986; Potvin *et al.*, 1991).

Muscle fibers sampled by the EMG surface electrodes were assumed to be representative of, and linearly related to the net muscle force. Lippold (1952) and Moritani and DeVries (1978) demonstrated linear relationships between surface EMG activity and voluntary isometric joint torque. Conversely, Zuniga and Simons (1969), Vredenbregt and Rau (1973), and Komi and Viitasalo (1976) measured EMG proportional to the square or the isometric joint torque. There is, clearly disagreement as to whether EMG is linearly or non-linearly related to force. Hof and Van Den Berg (1977) suggest EMG is linearly related to muscle force, whereas measured non-linear relations result from the influence of muscle co-contraction upon joint torque. EMG-assisted models account for muscle coactivity, and therefore a linear EMG–force relation was most appropriate. Physiologically valid muscle stress and high R^2 values were generated by the model independent of the weight of the lifted load (Table 1). Thus, assumption of a linear EMG–force relationship provided an excellent model output while avoiding overly complex biomechanics.

Although the literature warns of inherent interpretive difficulties when analyzing free-dynamic EMG, a distinction has been made here between ballistic and free-dynamic motion. The model is valid only for those motions wherein the time delay between the onset of myoelectric activity and muscular contractile force is minimal, i.e. smooth lifting motions. Redfern (1992) noted that during ballistic motion, there is often an initial burst in EMG activity not immediately reflected in force output. The Hill (1938) model of muscle contraction incorporating a contraction element and a series elastic element poorly predicts this type of viscous phenomena (Zahalak, 1986). A lifting model simulating only smooth trunk motions benefits by avoiding this poorly understood area of muscle mechanics. Future efforts should be dedicated to developing biomechanics analyses capable of including ballistic style lifting task associated with muscle activity bursts.

Spinal loads predicted by the free-dynamic, EMG-assisted model represent the three-dimensional forces in the lumbar region of the spine. Without an accurate measure of the dynamic orientation of individual vertebrae, biomechanical loads on specific elements of the spine cannot be modeled with this analysis. By including a geometric model of the lumbar spine dynamics, future efforts may employ the results from an EMG-assisted model to determine multi-dimensional load profiles on individual spinal elements.

These analyses have demonstrated that EMG-assisted modeling techniques may be employed to assess the biomechanical influence of trunk muscle coactivity upon spinal loads that occur during realistic representations of lifting tasks without the need for muscle activity approximations. The benefit of an EMG-assisted, free-dynamic, lifting model is the insight that it can provide into the effects of motion induced, muscle co-activity on spinal loading. Thus, these techniques may be employed to investigate the etiology of low-back disorders and reduce the risk of occupational injury.

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