■ Effect of Electromyogram–Force Relationships and Method of Gain Estimation on the Predictions of an Electromyogram-Driven Model of Spinal Loading

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Study Design. An experimental study of fatiguing isometric trunk extension was conducted to investigate the spinal loading estimated from an electromyogram-assisted biomechanical model.

Objective. To evaluate the sensitivity of the model outputs to two crucial assumptions: electromyogram–force relationship and method of gain estimation.

Summary of Background Data. In the proposed electromyogram-assisted models of the trunk, the nature of the electromyogram–force relationship and the wide variation in reported muscle gains can result in a wide variation in estimates of spinal loading. Given the absence of any gold standard for validation of muscle forces, the delineation of confidence intervals for the estimated loads has become critical.

Methods. Ten subjects performed a fatiguing isometric trunk extension while the net muscular torque output and trunk muscular activity were measured. An electromyogram-assisted model was used to estimate the torque output and spinal loading. Linear and nonlinear erector spinae electromyogram–force relationships and three methods for gain estimation were investigated: constant gain determined from an initial maximum extension exertion, constant gain based on the fatiguing exertion, and a time-varying gain from the fatigue test.

Results. The predicted torque was not sensitive to the electromyogram–force relationship; the nonlinear model produced 10% lower estimates of peak spinal compression force and 14% higher estimates of peak anterior shear force. The gain determined from an initial calibration exertion underestimated the external torque and underpredicted the peak compression force by 20%, compared with gains calculated in the fatigue test.

Conclusion. The nature of the electromyogram–force relationship and of the method for estimating the gain significantly affect the outcomes of an electromyogram-assisted model of spinal loading. [Key words: electromyogram-assisted model, electromyogram-force, isometric trunk extension, lifting, spinal loading] Spine 1998;23:423–429

In the past 15 years in many models of spinal loading, attempts have been made to quantify how much force the spine experiences during industrial work tasks.[3,4,8,9,16,24,29] However, estimated spinal loading has best been used as a relative index of severity, because of the inability to validate the predicted values by direct measurement of forces in the spine. As the fidelity of the models improves, the comparison of predicted loading with tissue tolerances may provide insight into injury mechanisms.[14] Therefore, it is important to identify to which assumptions the output of the model is most sensitive. For example, Nussbaum et al have demonstrated that the geometric configuration of the muscles affects the predicted muscle forces in optimization-based models.[19]

Electromyogram-assisted models of spinal loading provide benefits over their purely optimization-based counterparts because they are not limited by the constraints of the optimization–cost function and are sensitive to inter- and intrasubject variability in muscle recruitment.[9] A significant assumption, which may alter the magnitude of the predicted forces when using electromyogram-assisted models, is the nature of the electromyogram–force relationship. Because muscle tensions cannot be measured by noninvasive means, it is customary to derive the electromyogram–torque relationship from the net joint torque measurement. Both the linear and nonlinear electromyogram–torque relationship have been documented for the erector spinae.[1,2,23,26-28] It is important to stress that the nonlinearity of the electromyogram–torque correlation may be related to the

recruitment strategy within the multiple muscle system and not necessarily to the electromyogram–force correlation of the muscle.[12,22] However, the existence of nonlinear electromyogram–force correlations have been observed in muscles that contain approximately equal proportions of fast-twitch and slow-twitch fibers.[30] The presence of a nonlinear correlation may result in decreased accuracy in estimating torque generation and spinal loading, if linear models are used.

The invariance of the maximum stress that can be developed in muscle during full activation (hereafter referred to as *gain*) is another assumption that requires validation. The electromyogram-assisted,[9,16] optimization,[8,24] and hybrid models[3,4] of spinal loading all use the maximum stress-generating capability of muscle as an input into the models. Some researchers have assumed a given value of the maximum stress,[17] whereas others estimate it on the basis of various criteria to match the measured external moments with the predicted internal moments.[9,29] The assumption of constant gain (input or estimated) for muscle is likely to be violated when the muscle is fatigued, because of reduction in force output, given equivalent activation. Therefore, researchers have been careful not to apply electromyogram-assisted models to these situations. The method for estimating the gain, because of its influence on the magnitude of the estimated loading, is critical if the latter estimates are ever to be compared with the tissue tolerances.

Consequently, the purpose of the current study was to investigate the effect of the electromyogram–force and constant gain assumptions on the magnitude of spinal loading and the error in estimating the torque. Subjects performed a fatiguing isometric trunk extension while the trunk torque output and muscle activity were measured. Using variations of an electromyogram-assisted model of the spine, lumbosacral compression and shear forces were estimated. It was hypothesized that alterations in the electromyogram–force correlation and gain assumptions would result in significantly different errors in the prediction of torque production and significantly different predicted spinal compression and shear forces.

■ **Methods**

Ten male subjects (25.8 years old, range 20–32 years; mass, 83.5 ± 10.5 kg; height, 183 ± 6 cm) volunteered for the study. They were cleared for participation after a physical screening by an orthopedic surgeon. The subjects then read and signed the informed consent approved by the institutional review board of The Ohio State University. The subjects were instrumented with 10 pairs of bipolar surface electrodes (In Vivo Metric, Healdsburg, CA) 3 cm apart, to monitor the electromyographic activity bilaterally of the latissimus dorsi, lumbar erector spinae, rectus abdomini, external obliques, and internal obliques in the locations specified by Mirka.[18] To prepare for electrode placement, hair was shaved from the site and the surface was abraded with a cotton ball soaked with an alcohol solution. A KIN-COM 500H (Chattecx, Chattanooga, TN) dynamometer was used to provide isometric resistance to the

Figure 1. Isometric endurance test experimental setup.

subjects as they performed the endurance test. Each subject extended against a bar placed across his back, 28 cm vertically from the approximate location of his lumbosacral (L5–S1) disc. The bar was fixed to a load cell that measured the force exerted by the subjects (Figure 1).

Subjects were placed in the frame of the dynamometer. The trunk was flexed 5° from vertical (measured using an inclinometer), because of the difficulty in generating extensor torque in the upright (0°) position.[21] With legs and pelvis restrained and trunk flexed 5° from vertical, the subjects performed isometric maximum voluntary contractions (MVCs) of 5 seconds' duration, in trunk flexion, extension, right and left lateral bending, and right and left axial rotation. Resistance in lateral bending and axial rotation was provided by Velcro straps. The MVCs were separated by 2 minutes' rest to avoid fatigue.

The endurance test was designed to cause trunk muscle fatigue to investigate the ramifications on the constant gain assumption. Therefore, subjects were requested to track a time-varying reference extension torque, which was displayed on a feedback computer monitor. The reference torque was set at levels approximately equal to 55–95% (in increments of 10%) of the extension MVC, randomly within and between subjects. Because of the random nature of the setting, few of the subjects exerted at all increments of the reference torque. In fact, for those subjects who by chance received the higher levels initially, the endurance test may have lasted for only two or three changes in torque level. The program produced a bandwidth consisting of the reference torque, as a percentage of the subject's extension MVC, ±5% of the MVC. The reference bandwidth and the measured torque were plotted simultaneously on the output monitor, so that the subject was able to maintain his torque level within the reference bandwidth. Each reference torque lasted from 3 to 10 seconds (in increments of 1), randomly within and between subjects. Thus, each subject was unaware of the change in reference torque that would occur, as well as when it would occur. The test continued until the subject was no longer able to maintain the reference torque for two consecutive changes in torque. During the following analysis, the endurance time was objectively defined as the point when at least half of the generated torque values within a 2-second moving window were outside a tolerance of ±10%. An example of the endurance test, demonstrating the reference and measured torque output of the subject, is shown in Figure 2. A more complete description of the endurance test is presented in Sparto et al.[25]

The electromyography signals were preamplified by a factor of 1000, band-pass filtered at 12 and 1000 Hz, and further

amplified to increase the resolution of each muscle's signal. The analog voltage signals from the load cell and electromyogram system were digitized at 1000 Hz by a Data Translation (Marlboro, MA) 12-bit A/D board and stored on a personal computer. The digitization rate was determined assuming the highest frequency component in the signal to be 500 Hz. The electromyogram and measured torque data from the endurance test were input into an electromyogram-assisted biomechanical model of the spine to estimate the loading of the spine in the following manner.[9] The model assumes that the force $F_i(t)$ developed in the ith muscle is equal to the product of the normalized electromyogram $NEMG_i(t)$, physiologic cross-sectional area $PCSA_i$, length–tension $f(l_i)$ and velocity–tension $g(v_i)$ modulating factors, and the gain $G_i(t)$:

$$F_i(t) = NEMG_i(t) \cdot PCSA_i \cdot f(l_i \cdot g(v_i) \cdot G_i(t) \qquad (1)$$

The electromyogram from the 10 trunk muscles was rectified and a root-mean-square (rms) value was generated by a 1-second moving window. The processed electromyogram curves were then normalized by the maximum electromyogram obtained for the respective muscle from any of the MVCs during the calibration trials. The baseline electromyogram was verified to be negligible in this slightly flexed position. The PCSAs of the muscles were scaled relative to the trunk width and depth of the subjects.[9] The velocity–tension modulating factor was set at 1, because the test was isometric. Furthermore, the length–tension factor was set at 1, because the resting length of the muscles was assumed to occur when the torso was upright[20]; the effect of 5° of forward flexion on the length–tension factor was considered to be negligible. Because the equation still has two unknowns, $F_i(t)$ and $G_i(t)$, another equation is needed for the system to become determinate at each instant of time.

The second equation comes from the equilibrium condition. To satisfy the equilibrium constraint, the measured external torque, $t_{meas}(t)$, about the lumbosacral disc is balanced by the internal moment generated by the 10 muscles:

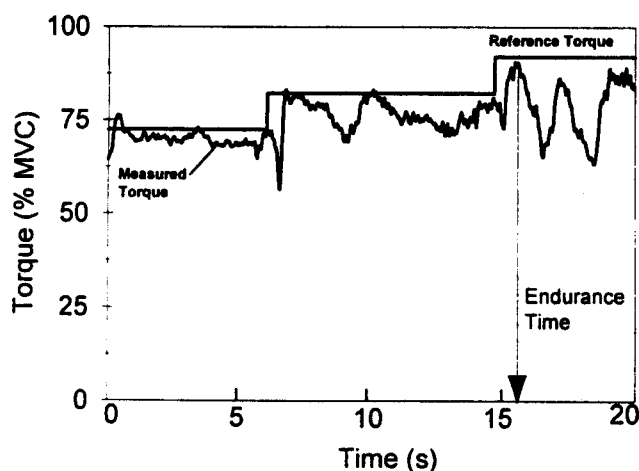$$\vec{\tau}_{pred}(t) = \sum_{i=1}^{10} \vec{r}_i \times \vec{F}_i(t) \qquad (2)$$
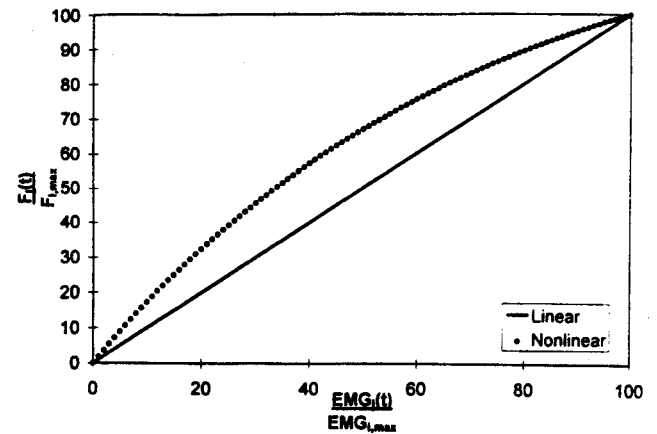

Figure 3. Linear and nonlinear electromyogram–force correlations for the erector spinae muscles. The nonlinear correlation was determined using equation 3. (Adapted from Potvin[23])

where r_i is the moment arm from the center of the lumbosacral disc to the ith muscle. The moment arms and muscle lines of action are also from Granata and Marras.[9] It is recognized that the muscle moment arms and lines of action significantly affect the predicted spinal loads.[19] After making simplifying assumptions that the nonsagittal plane torque is negligible[21] and that the gain is equal for all muscles,[9] the instantaneous force and gain can be predicted by solving equations 1 and 2 simultaneously. The compression and shear joint reaction forces are then computed by summing the externally applied loads (body weight and shear, depending on load application) and internally estimated muscle forces in the three orthogonal directions aligned with the orientation of the lumbosacral disc.

Two model variations were implemented to determine the effect of the electromyogram–force correlation on the estimates of spinal loading. The linear and nonlinear electromyogram–force correlations are displayed in Figure 3. In the linear case, the relative amount of force $F_i(t)$ produced by the erector spinae was assumed to be proportional to that in the normalized electromyogram. In the nonlinear case, an exponential correlation between the normalized electromyogram and force $F_i(t)$ in the erector spinae was assumed, according to the experimental data of Potvin[23]:

$$\frac{F_i(t)}{F_{i,max}} = \left[\frac{e^{-1.4 \cdot \frac{EMG_i(t)}{EMG_{i,max}}} - 1}{e^{-1.4} - 1} \right] \qquad (3)$$

The nonlinear correlation was implemented for the erector spinae muscles only, because of the lack of experimental data for the other muscles.

The value of gain was estimated by three methods. In the first method, the value of gain G_{ls_max} minimized the sum of squared error between the externally measured and internally predicted extension torques computed during the maximum extension torque calibration trial conducted at the start of the testing. This least-squares method reduces the error variance and bias in estimating the gain, compared with the error in the method of taking an average of the instantaneous gain:

$$\min \sum_{j=1}^{N} (\tau_{meas} - \tau_{pred})^2 \qquad (4)$$

Figure 2. Example of an isometric endurance test, indicating the reference and measured torque.

Table 1. Descriptive Statistics and Results From the Analysis of Variance (ANOVA) Showing the Effect of the EMG–Force Relationship on the G_{ls_end}, rms and Peak Relative Error Between the Measured and Predicted Sagittal Torque (ERR), and the rms and Peak Predicted Compression Force (COMP) and Anterior Shear Forces (SHEAR) at the Lumbosacral Joint

Dependent Variable	EMG–Force Relationship		Results of ANOVA (P value)
	Nonlinear [mean (SD)]	Linear [mean (SD)]	
G_{ls_end} (N/cm²)	30 (9)	43 (13)	<0.0035
RMS ERR (%)	14 (3)	16 (4)	<0.0035
Peak ERR (%)	41 (13)	42 (12)	NS
RMS COMP (N)	1752 (286)	1883 (307)	<0.0035
Peak COMP (N)	2666 (408)	2969 (504)	<0.0035
RMS SHEAR (N)	786 (166)	692 (193)	<0.0035
Peak SHEAR (N)	1067 (275)	950 (329)	<0.0035

NS = not significant.

where j is a discrete sample index, N the total number of discrete samples, τ_{meas} the externally measured extension torque, and τ_{pred} the internally predicted torque from equation 2. The second method computed the least squares gain for the duration of the endurance test G_{ls_end}. For the third estimation technique, the gain remained a function of time G_{local}, so that the internal torque equaled the external torque at each instant. The linear electromyogram–force correlation was used in the model for all the gain variations, and the G_{ls_end} was used in the model to test the effect of the electromyogram–force correlation.

For each of these model variations, the rms and peak relative error between the measured and predicted torque, during the entire endurance test, were calculated to investigate the effect of the model variations on the accuracy of the torque estimates. Similarly, the rms and peak predicted anterior shear and compression forces were determined, to quantify the effect of these variations on spinal loading. The effects of the type of electromyogram–force correlation and method of gain estimation on the errors in torque prediction and estimates of compression and shear forces were tested using one-way analysis of variance with repeated measures. Posthoc analysis of the significant effects was performed using Tukey's test. Because of the large number of comparisons, a Bonferroni correction was applied to the overall Type I error rate of 0.05; thus, a was set to 0.0035.

■ **Results**

The mean ISD endurance time for the test was 38 ± 27 seconds. The randomization of the torque levels within and between subjects was confirmed by analysis of variance, with results demonstrating that there was no systematic increase or decrease in the reference torque level as the test progressed. The effect of the electromyogram–force correlation was significant ($P < 0.0035$) for all the dependent measures except the peak relative error between the measured and predicted torque (Table 1). As a note, the estimation of the gain was approximately 30% less using the nonlinear electromyogram–force correlation ($P < 0.0035$), because more erector spinae force is

produced per unit of electromyogram using this correlation, except at the highest levels of exertion (100% MVC). On average, the model underpredicted the output torque; however, in some cases, the peak error in overprediction was as great as the peak error in underprediction. The rms relative error between the predicted extension torque and measured extension torque was significantly greater for the linear electromyogram–force correlation. The mean ±SD rms errors were 13.2 ± 4.2 Nm and 11.8 ± 3.4 Nm for the linear and nonlinear correlation, respectively, and the maximum errors were 27.3 ± 9.7 Nm and 24.6 ± 8.8 Nm. The linear electromyogram–force correlation resulted in greater predicted compression forces and lower anterior shear forces ($P < 0.0035$).

The effect of method of gain estimation resulted in significantly different predictions of torque production and spinal loading (Table 2). The errors in estimating the torque while using a constant gain estimated from the initial maximum extension exertion could be substantial during a fatiguing isometric exertion. For example, the peak error using this method was 42.1 ± 8.6 Nm. Furthermore, the rms relative error between the measured and predicted torque using the G_{ls_max} method was twice that in the G_{ls_end} condition. The compression and shear predictions using the maximum extension exertion method were consistently lower than were those in the other methods. Generally the two other methods produced similar estimates of spinal loading, except for the lower peak compression force predicted by the G_{local} method.

■ **Discussion**

The model performed slightly better in predicting the external torque while assuming a nonlinear erector spinae electromyogram–force correlation (Table 1). The predicted compression force was less in the nonlinear condition compared with that in the linear condition,

Table 2. Descriptive Statistics and Results From the Analysis of Variance (ANOVA) Showing the Effect of the Gain Estimation Method on the Gain, rms and Peak Relative Error Between the Measured and Predicted Sagittal Torque (ERR), and the rms and Peak Predicted Compression Force (COMP) and Anterior Shear Forces (SHEAR) at the Lumbosacral Joint

Dependent Variable	Method of Gain Estimation			Results of ANOVA (P value)
	G_{ls_end}	G_{ls_max}	G_{local}	
Gain (N/cm²)	43 (13)	32 (8)	NA*	<0.0035
RMS ERR (%)	16 (4)	33 (10)	0 (0)†	<0.0035
Peak ERR (%)	42 (12)	51 (9)	0 (0)†	<0.0035
RMS COMP (N)	1883 (307)	1509 (405)	1939 (315)	<0.0035
Peak COMP (N)	2969 (504)	2334 (657)	2341 (369)	<0.0035
RMS SHEAR (N)	692 (193)	634 (154)	711 (198)	<0.0035
Peak SHEAR (N)	950 (329)	841 (260)	964 (353)	<0.0035

* Population means are not available for this measure because it was allowed to vary as a function of time.
† The error degenerates to 0, based on the definition of the method.

because more erector spinae force per unit of electromyogram was generated, resulting in greater extensor moment and less gain. Consequently, the lesser gain produced lower force estimates in the muscles other than the erector spinae, which resulted in a reduction in compression forces relative to the linear condition. Those investigators who intend to predict the external moments based on the electromyogram activities are less influenced by the nature of the electromyogram–force correlation assumed in the model,[28] whereas those who intend to estimate the spinal loading are significantly affected, as shown in the current results. Although nonlinear (bilinear) correlation between trunk extensor electromyogram and torque have been observed,[26,27] Stokes et al[26] recommended that a linear fit was the best model for any subject chosen at random. Given the kinetic redundancies (more muscles than those required for static equilibrium) and the presence of coactivation, the electromyogram–force correlation of individual trunk muscles cannot be determined on the basis of electromyogram–trunk torque correlation. The reference torques were set at high levels of MVC to fatigue the subjects in a short time. Therefore, the effect of the electromyogram–force correlation was not investigated over the full range. Future studies that incorporate the entire range are needed to extend the sensitivity of the model to this effect for exertions below 55% MVC.

Similarly, the evaluation of the best method for determining the gain was equally difficult. Using G_{ls_max}, the relative error between the measured and predicted torque was twice as much as that recorded when G_{ls_end} was used. In addition, because G_{ls_max} was less, estimates of spinal loading were less than those recorded when the other two methods were used. The determination of gain is a critical step for obtaining realistic estimates of spinal loading. A set of complex triaxial calibration tasks is needed for the proper system identification and computation of generalizable gain estimates. Thelen et al[29] showed that inclusion or exclusion of axial exertions in the calibration sets significantly affected the accuracy of their model parameters. Furthermore, inclusion of a diverse set of calibration tasks was suggested as an experimental remedy to the problem of multi-colinearity. The colinearity is caused by the high level of correlation in the activation patterns of muscles in synergistic groups, which leads to the higher bias and variance inflation in parameter estimations. Principal component regression has been suggested as a solution by Hughes et al,[13] and Thelen et al[28] have shown its limitations as well.

The poor performance of the model using the G_{ls_max} indicates that the gain values obtained from the maximum calibration exertions at the beginning of the testing day may not be accurate if subjects are expected to fatigue during the testing session. Therefore, such factors as fatigue that influence the maximum muscle force-generating ability must be considered in the development of the concept of gain. In this regard, the temporal pattern of G_{local} allows consideration of a fatigue-
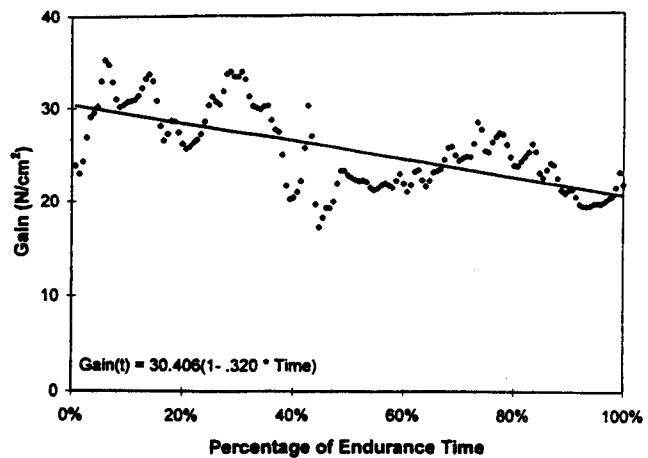

Figure 4. Variation in G_{local} as a function of fatigue during the same endurance test as that depicted in Figure 2. The linear function describing the correlation between G_{local} and percentage of endurance time, Gain(t) = 30.406 (1−0.320·Time), is shown. The statistics for the regression are $r^2 = 0.43$; $P < 0.05$.

modulating factor that may be used to extend the use of electromyogram-assisted models in fatiguing conditions. For example, in Figure 4, the variation in G_{local} through the duration of an endurance test is demonstrated (the same test shown in Figure 2). It is evident that even for relatively short durations of constant torque, there is wide variability in the gain, indicating the inherent difficulty in mapping 10 variable electromyogram signals to a single measured external torque. For this test, the least-squares estimate of gain was 23.9 and the mean estimate was 25.5. Superimposed on the plot is a linear function ($r^2 = 0.43$; $P < 0.005$) that was fit to the data. Therefore, the equation listed on the plot could be used as a modulation factor for this performance. Besides such empirical measures, other fatigue-modulating factors have been proposed, based on fiber type composition and biochemical measures.[7,11] Another less attractive alternative is to repeat the calibration trials frequently so the effect of fatigue can be accounted for. The effect of confounding factors (i.e., level of torque generation and changes in recruitment of alternate muscles),[25] especially if unsampled muscles become more prominent in torque generation (i.e., thoracic erector spinae or quadratus lumborum), may make the determination of the correlation between fatigue and gain difficult.

The determination of gain can be affected by several limiting factors. The assumptions regarding the length-tension modulation, velocity–tension modulation, and physiologic cross-sectional area all have a profound effect on the determination of the gain as formulated in this model. Although the range of the maximum stress for the single-fiber and motor unit preparations is relatively narrow (17–23 N/cm^2), the range reported in studies involving human performance is very large (11–70 N/cm^2), because of differences in fiber type proportions, muscular architecture, and functional anatomy.[6] Thelen et al[29] have used system identification techniques for estimating

different gain values for different functional sets of trunk muscles, thus relaxing the single gain assumption. Another practical issue is that trunk muscles with their complex functional arrangements may not elicit their maximum activities during any of the selected calibration tasks.[15] Furthermore, although the ultimate or yield stress of muscle is a musculotendon property, the muscle gain determined in the current experiment is dependent on the subject's ability to perform the maximum voluntary contractions; therefore, the gain is in part a psychophysical measure.[5]

It is also important to recognize that the findings about the response of gain apply only to these long-duration isometric exertions. The electromyogram-assisted model was primarily used to infer the changes in spinal loading caused by a fatiguing task, which represents a worst-case scenario for the model's performance because of the limited range of torque developed and the change in muscle activation related to fatigue. Application of the model to conditions for which the model was intended—short duration dynamic lifting exertions in which fatigue is unlikely—would not require adjustment of the gain. However, there is some evidence that for an extended session of repetitive lifting, the gain should be adjusted during the course of the session because of effects of muscle fatigue.[10]

In the absence of a technological breakthrough that would allow a better estimation of muscle force and maximum stress-generating capabilities, the electromyogram-assisted models of the trunk can be used to evaluate the spinal loads during physical exertions. However, the assumptions used in these models must be scrutinized more closely and a confidence interval for each of these point estimates must be provided. In the current study, the same electromyogram patterns are used in finding the variance in the model predictions, based on manipulation of the assumptions that provide the fundamental basis of electromyogram-assisted models. In future studies, further sensitivity analysis should be considered to delineate the precision and confidence about estimated spinal loading. This will surely contribute toward delineating injury mechanisms when comparing the spinal loads with limit values of respective passive and active elements of the spine.[14]

Acknowledgments

References

1. Andersson GB, Örtengren R, Herberts P. Quantitative electromyographic studies of back muscle activity related to posture and loading. Orthop Clin North Am 1977;8:85–96.

2. Chapman AE, Troup JD. Prolonged activity of lumbar erector spinae. An electromyographic and dynamometric study of the effect of training. Ann Phys Med 1970;10:262–9.

3. Cholewicki J, McGill SM, Norman RW. Comparison of muscle forces and joint load from an optimization and EMG assisted lumbar spine model: Towards development of a hybrid approach. J Biomech 1995;28:321–31.

4. Cholewicki J, McGill SM. EMG assisted optimization: A hybrid approach for estimating muscle forces in an indeterminate biomechanical model. J Biomech 1994;27:1287–9.

5. Enoka R, Fuglevand A. Neuromuscular basis of the maximum voluntary force capacity of muscle. In: Grabiner M, ed. Current Issues in Biomechanics. Champaign, IL: Human Kinetics Publishers, 1993:215–36.

6. Fukunaga T, Roy RR, Shellock FG, Hodgson JA, Edgerton VR. Specific tension of human plantar flexors and dorsiflexors. J Appl Physiol 1996;80:158–165.

7. Giat Y, Mizrahi J, Levy M. A musculotendon model of the fatigue profiles of paralyzed quadriceps muscle under FES. IEEE Trans Biomed Eng 1993;40:664–73.

8. Gracovetsky S, Farfan HF, Lamy C. The mechanism of the lumbar spine. Spine 1981;6:249–62.

9. Granata KP, Marras WS. An EMG-assisted model of loads on the lumbar spine during asymmetric trunk extensions. J Biomech 1993;26:1429–38.

10. Granata KP, Marras WS. Changes in trunk and spine biomechanics during repetitive fatiguing exertions. Proceedings of the Fifteenth Southern Biomedical Engineering Conference IEEE, Piscataway, NJ, PK Bajpai (ed), 1996;439–42.

11. Hawkins D, Hull ML. Muscle force as affected by fatigue: Mathematical model and experimental verification. J Biomech 1993;26:1117–28.

12. Hof AL, van den Berg J. Linearity between the weighted sum of the EMGs of the human triceps surae and the total torque. J Biomech 1977;10:529–39.

13. Hughes RE, Chaffin DB, Lavender SA, Andersson GBJ. Evaluation of muscle force prediction models of the lumbar trunk using surface electromyography. J Orthop Res 1994;12:689–98.

14. McGill SM. Estimation of force and extensor moment contributions of the disc and ligaments at L4–L5. Spine 1988;13:1395–402.

15. McGill SM. A myoelectrically based dynamic three-dimensional model to predict loads on lumbar spine tissues during lateral bending. J Biomech 1992;25:395–414.

16. McGill SM, Norman RW. Partitioning of the L4–L5 dynamic moment into disc, ligamentous, and muscular components during lifting. Spine 1986;11:666–78.

17. McGill SM, Norman RW. Effects of an anatomically detailed erector spinae model on L4/L5 disc compression and shear. J Biomech 1987;20:591–600.

18. Mirka GA. The quantification of EMG normalization error. Ergonomics 1991;34:343–52.

19. Nussbaum MA, Chaffin DB, Rechtien CJ. Muscle lines-of-action affect predicted forces in optimization-based spine muscle modeling. J Biomech 1995;28:401–9.

20. Nussbaum NA, Chaffin DB. Development and evaluation of a scalable and deformable geometric model of the human torso. Clin Biomech 1996;11:25–34.

21. Parnianpour M, Nordin M, Kahanovitz N, Frankel V. The triaxial coupling of torque generation of trunk muscles during isometric exertions and the effect of fatiguing isoinertial movements on the motor output and movement patterns. Spine 1988;13:982–92.

22. Perry J, Bekey GA. EMG–force relationships in skeletal muscle (review). Crit Rev Biomed Eng 1981;7:1–22.

23. Potvin JR. The influences of fatigue on hypothesized mechanisms of injury to the low back during repetitive lifting. University of Waterloo, Ontario, Canada, 1992. PhD Thesis.

24. Schultz A, Andersson G, Ortengren R, Haderspeck K, Nachemson A. Loads on the lumbar spine. Validation of a biomechanical analysis by measurements of intradiscal pressures and myoelectric signals. J Bone Joint Surg [Am] 1982;64:713–20.

25. Sparto PJ, Parnianpour M, Marras WS, Granata KP, Reinsel TE, Simon SR. Neuromuscular trunk performance and spinal loading during a fatiguing isometric trunk extension with varying torque requirements. J Spinal Disorders 1997;10:145–56.

26. Stokes IA, Rush S, Moffroid M, Johnson GB, Haugh LD. Trunk extensor EMG–torque relationship. Spine 1987;12:770–6.

27. Tan JC, Parnianpour M, Nordin M, Hofer H, Willems B. Isometric maximal and submaximal trunk extension at different flexed positions in standing. Triaxial torque output and EMG. Spine 1993;18:2480–90.

28. Thelen DG, Schultz AB, Ashton–Miller JA. Quantitative interpretation of lumbar muscle myoelectric signals during rapid cyclic attempted trunk flexions and extensions. J Biomech 1994a;27:157–67.

29. Thelen DG, Schultz AB, Fassois SD, Ashton–Miller JA. Identification of dynamic myoelectric signal-to-force models during isometric lumbar muscle contractions. J Biomech 1994b;27:907–19.

30. Woods JJ, Bigland–Ritchie B. Linear and non-linear surface EMG/force relationships in human muscles. An anatomical/functional argument for the existence of both. Am J Phys Med 1983;62:287–99.

■ Point of View

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The authors should be commended for their efforts to identify potential sources of error in electromyogram-based models used to estimate trunk loading. Recently, an increased emphasis has been placed on studying the effects of muscle fatigue on spine mechanics. The current results indicate that large errors will result if a single value for maximum muscle stress (termed gain) is used throughout a fatiguing contraction. However, most electromyogram-based trunk models already use some method to manipulate the muscle gain such that the predicted net moment is always equal to the measured moment. In this way, these models currently account for the time-dependent decreases in muscle gains that may be expected under fatiguing conditions. However, one limitation of most models is that each muscle is assumed to have the same gain at any instant in time. For example, it is not likely that the gain of the abdominal muscles would have been the same as that of the more fatigued erector spinae muscles at the end of a trunk extensor endurance trial used in this study.

The decreasing gain illustrated in Figure 4 may represent the decreasing force-generating capacity of the fatiguing trunk muscles. However, it appears that the gain was also dependent on the instantaneous torque level. There appears to be an immediate overall decrease in gain after the target torque increased from approximately 75% to 85% of maximum (Figures 2 and 4). Whereas fatigue would be expected to result in gradual gain changes, this rapid force-related change may have resulted from the use of a linear normalization. The force associated with a given submaximal electromyographic level would be higher with a nonlinear normalization (compared with linear), and the magnitude of this difference decreases to zero as the electromyogram increases from the mid range to maximum activation (see Figure 3). Therefore, as force was increased, the linear normalization would underestimate force less, and the gain could decrease. In a protocol involving changing force levels, the use of the nonlinear normalization may have more effectively isolated the relation between gain and fatigue. Similarly, it may have been easier to isolate the effects of normalization linearity during rested contractions without the confounding effects of progressive fatigue. However, regardless of these issues, Sparto et al have successfully demonstrated the sensitivity of electromyogram-based trunk model outputs to assumptions of gain and the nature of the electromyogram–force relation. Future work is needed to identify the most appropriate assumptions to optimize the accuracy of such models.