

## Neuromuscular Trunk Performance and Spinal Loading During a Fatiguing Isometric Trunk Extension with Varying Torque Requirements

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**Summary:** A novel testing protocol was used to investigate changes in neuromuscular performance, muscle recruitment, and spinal loading as subjects became fatigued while performing an isometric endurance test of varying torque requirements. There was decreased accuracy in maintenance of a reference torque but no change in response time as subjects became fatigued. The study of trunk-muscle recruitment indicated significant increases in internal oblique and latissimus dorsi muscle activity. This change in recruitment led to changes in spinal loading despite a relatively constant torque output. The use of an electromyogram (EMG)-assisted model demonstrated that when subjects are expected to become fatigued during test performance, the assumption of a constant maximal stress capacity of the muscle may not be robust. **Key Words:** Isometric trunk extension—Spinal loading—Accuracy—Performance—Trunk-muscle recruitment.

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Trunk muscle endurance appears to be a better indicator of future low back pain than does strength (3,4,25,26,32,66). For this reason, researchers have increasingly investigated trunk-muscle endurance by using various testing protocols, ranging from isometric to freely dynamic repetitive lifting. However, for clinical use, the predictive value of trunk-muscle endurance tests must be established. Much of the previous research focused on physiologic changes, such as efficiency, power output, heart rate, and  $VO_{2MAX}$ , that result from fatigue (21,22,31,51,71). Other research investigated the effect of fatigue on biomechanical parameters, such as external torque about the lumbar intervertebral discs, in addition to compression and shear forces on the discs (52,69). Despite this expansive body of research, the effect of fatigue on

the quality of the neuromuscular performance of the trunk has been lacking.

Parnianpour et al. (49) instructed subjects to flex and extend from the upright position to their peak trunk flexion as quickly and accurately as possible against a resistance of 70% of their maximal voluntary exertion. A reduction in performance occurred at the end of the task, indicated by decreased sagittal trunk range of motion and velocity, whereas the coronal and transverse plane range of motion significantly increased. The biomechanical implications of the fatigue-induced performance decrement remain unclear, but it is hypothesized that if muscles are required to respond to an unexpected load or a demand that is greater than the compromised muscles' capability, the brunt of the load may be shifted to the passive tissues, and possible injury may result (49,55,58). Although the decline in torque output with fatigue is universally recognized, how fast the trunk muscles are able to respond to a change in torque demand as the muscle becomes fatigued is unknown. Fatigue affects the rate of

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Received May 1, 1996; accepted September 17, 1996.  
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force development and that of force relaxation (5,23). It is hypothesized that if the neuromuscular response caused by fatigue is inappropriate or delayed, a more clear injury mechanism may be identified. A testing paradigm is designed to test this aspect of the performance of the trunk musculature.

The investigation of the changes in the recruitment patterns and spectral characteristics of the electromyogram (EMG) is necessary to understand how the muscles coordinate their many degrees of freedom to maintain a required torque output. For instance, Dieen et al. (14) described alternating activity in the trunk extensors that resulted in an increase in endurance during an isometric extension. Similarly, evidence obtained from the frequency spectrum of the EMG showed that motor-unit firing frequencies decline during fatigue (5), probably to optimize force production, given the slowing of the relaxation rate of muscle (6). Although these analyses are useful for developing hypotheses of how the muscles are able to generate constant torque despite a decline in the maximal force-generation capacity of each muscle (38), they provide no understanding of how these changes affect the spinal loading and thus injury mechanisms.

The consequences of the alteration in muscle-recruitment patterns caused by fatigue on the internal spinal loading have been investigated by using EMG-assisted models. The motivation for using these models is that they provide a reduction in the parameters (one external torque and 10 EMG measurements to three joint reaction forces) and they produce quantities that can be compared to strength-tolerance data for spinal tissues (60,72). EMG-assisted (19,40), optimization (18,57), and hybrid models (10,11) of spinal loading all use the concept of the maximal stress-generating capability of muscle. The question that faces all researchers is how the change in the capability of the contractile apparatus to generate tension, resulting from fatigue, affects the maximal stress-generating capability of muscle. In other words, should models incorporate a fatigue-modulating factor in their models? The use of fatigue-modulating factors based on percentage of muscle fiber type and metabolic factors have been validated (17,24). Because the magnitude of the muscle tension is directly proportional to the maximum stress-generating capability of muscle, the proper determination of the maximum stress is critical for obtaining reliable estimates of the spinal loading.

The purpose of our study was to discover the effect of fatigue on the quality of neuromuscular performance, muscle recruitment, and spinal loading during an isometric trunk extension at varying levels of exertion. An isometric trunk-extension test was designed to measure the capability of the trunk musculature to respond to unexpected changes in torque demand while the muscles were becoming fatigued. In addition, the trunk-muscle recruitment

patterns and frequency content of the EMG were measured to understand how the neuromuscular system maintains a given torque level during fatigue. Subsequently an EMG-assisted model of the spine was used to examine the effects of the muscle recruitment on the spinal loading and to discern, by incorporation of a fatigue-modulating factor, the effect of fatigue on the spinal loading.

## MATERIALS AND METHODS

Ten men [mean age, 25.8 years; range, 20–32 years; weight, 83.5 kg (SD, 10.5 kg); and height, 183 cm (SD, 6 cm)] volunteered for the study. Before participating, they were cleared to participate after a physical screening by an orthopaedic surgeon. The subjects then read and signed the informed consent, approved by the Biomedical Sciences Committee, Research Involving Human Subjects Committee of The Ohio State University.

### Isometric Endurance Test Design

The KIN-COM 500H (Chattecx, Inc., Chattanooga, TN, U.S.A.) dynamometer can provide computer-controlled isometric and isokinetic resistance. It was modified with hardware designed in the Biodynamics Laboratory so that the trunk could be tested in various positions of flexion and asymmetry (33–35,41,42). It was used to provide isometric resistance to the subjects as they performed the endurance test. Each subject extended against a bar placed across his back, 28 cm vertically from his lumbosacral (L5/S1) disc. The bar was fixed to two load cells, which were attached to the frame of the dynamometer. The load cells measured the force exerted by the subjects. Figure 1 shows the experimental setup.

The experiment was designed to cause trunk extensor-muscle fatigue while subjects were requested to track a time-varying reference extension torque. For safety reasons, it was decided not to use eccentric action in the muscle, which has been implicated as an injury mechanism (46). Consequently, a load change was brought about by introducing an unexpected change in the cognitive task demand (i.e., the reference extension torque that the subjects were instructed to generate). The subjects were instructed to track the reference torque by responding to the change in task demand as quickly as possible. The software visually provided, in real time, the reference level of torque required of the subject and the actual measured trunk-extension torque generated by the subjects.

## PROCEDURE

The subjects were instrumented with 10 pairs of bipolar surface electrodes (In Vivo Metric, Healdsburg, CA,

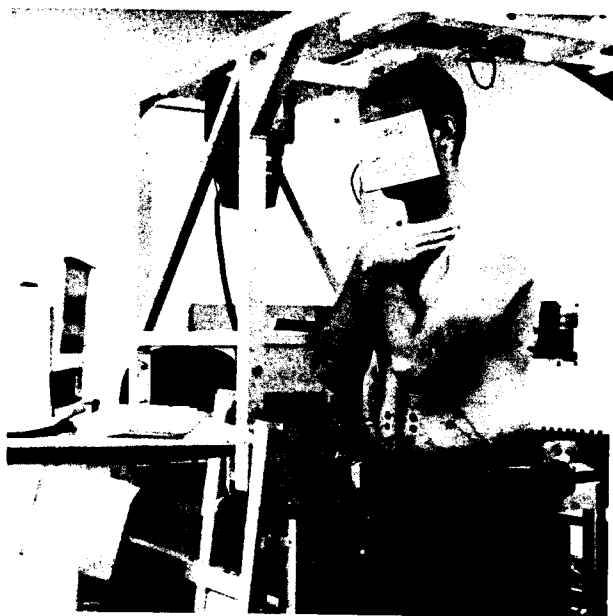


FIG. 1. Isometric endurance test experimental setup.

U.S.A.), 3 cm apart, to monitor the EMG activity bilaterally of the latissimus dorsi, erector spinae, rectus abdomini, external obliques, and internal obliques. The electrodes were placed in the following locations, per Mirka (41). For the rectus abdominis, 3 cm lateral and 2 cm superior to the umbilicus; external oblique, 10 cm lateral to the umbilicus at an angle of 45°; internal oblique, 10 cm lateral to the midline, just superior to the iliac crest and inferior to the latissimus dorsi ("lumbar triangle"), at an angle of 45°; latissimus dorsi, lateral to T9 spinous process over the muscle belly; and erector spinae, 4 cm lateral to L3 spinous process. To prepare, hair was shaved from the site and the surface abraded by an alcohol solution.

Subjects were placed in the frame of the dynamometer. With legs and pelvis restrained by a frame and trunk flexed 5° from vertical, the subjects performed a series of maximal voluntary isometric contractions of 5-s 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 min of rest to avoid muscular fatigue (8). Only one contraction was used to establish the MVC in extension. In another study with the same group of motivated subjects, there was no difference in the between-day performance of maximal strength tests (64). Consequently it was assumed that one exertion was sufficient to provide a reliable estimate of the MVC. Furthermore, Smidt et al. (63) demonstrated a high intraclass correlation coefficient ( $r = 0.91$ ) for subjects performing three MVCs in the same position.

To become accustomed to the testing procedure and software, subjects performed a 45-s warmup exertion at submaximal target levels from 25 to 50% of the MVC. After another 2 min of rest, the endurance testing began. The peak torque value of the MVC in trunk extension was input into the reference torque-generation software. The reference torque was set at levels 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 of the reference torque, few of the subjects exerted at all increments of the reference torque. In fact, for those subjects that, 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,  $\pm 5\%$  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 s (in increments of 1) randomly within and between subjects. Thus each subject was unaware of the change in reference and 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. An example of the test for one subject is shown in Fig. 2.

#### DATA ANALYSIS

The EMG signals were preamplified with a gain of 1,000, band-pass filtered at 15 and 1,000 Hz, and further amplified with variable gain, by using visual identification to increase the resolution of each muscle's signal to the maximal range collected by the analog-to-digital (A/D) system. The analog voltage signals from the KIN-COM and EMG system were digitized at 1,000 Hz by a Scientific Solutions (Solon, OH, U.S.A.) 12-bit A/D board and stored on a Compaq Portable 386 computer.

The data from each endurance test were analyzed to find the endurance time. A program was written to determine whether the generated torque values were within a 10% tolerance of the reference torque. The endurance time was objectively defined as the point at which at least half of the generated torque values within a 2-s moving window were outside of the tolerance. The transition periods when a change in torque occurred were not considered in the analysis.

A primary goal was to determine whether the relative accuracy in maintaining the reference torque changed during the endurance test. The reference torque and measured torque data were subsampled at 100 Hz and converted into percentages of the peak torque from the extension MVC. The difference between the reference torque and the mea-

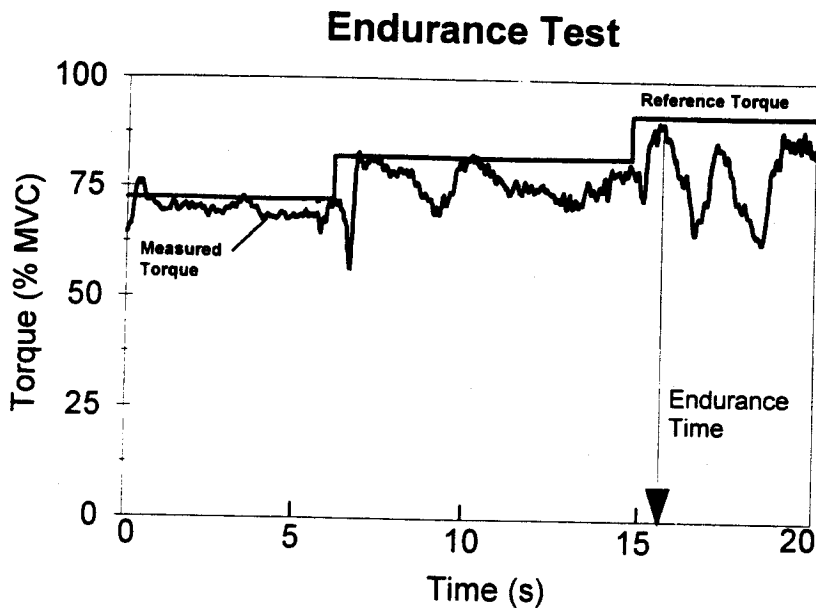


FIG. 2. Example of an isometric endurance test, indicating the reference and measured torque.

sured torque produced the relative error. For each second of data, a root-mean-square (RMS) value of the relative error was calculated ( $ERR_{rms}$ ), as well as the standard deviation of the relative error ( $ERR_{std}$ ).

Changes in the EMG content (temporal and spectral) of the muscle activation provide some evidence for the existence of fatigue, even when there is no change in muscle force. Because others have shown changes in muscle activation over time (13,53), it was attempted to discern whether there was any change in muscle recruitment during the endurance test. Therefore 1-s RMS values were determined for the time-domain EMG of the erector spinae, latissimus dorsi, and internal obliques. The median frequency of the power spectrum generated by the EMG signal is sensitive to fatigue-related changes during constant force, constant length contractions (12,15,28,38,43). Consequently it was desired to determine whether the median frequency parameter was sensitive to the time-varying levels of isometric exertions as well. Because the attempted task was trunk extension, it was hypothesized that greater fatigue would be present only in the extensor muscles (i.e., erector spinae, latissimus dorsi, and internal obliques); thus the spectral changes for these muscles only were reported.

The spectral analysis was performed in the following manner. Each second of data was split into quartiles, and the raw EMG from each 250-ms period was Hanning windowed and transformed into the frequency domain by using a 256-point Fast Fourier Transform with zero padding (Matlab; Mathworks, Natick, MA, U.S.A.). Subsequent multiplication of the Fourier transform by its complex conjugate produced the power spectral density function (PSDF). The four PSDFs from each second were

averaged to attenuate the random noise in the PSDF (28). The median frequency was calculated by finding the frequency at which the area under the PSDF is halved. The response of the median frequency (MF, in Hertz) to time (Time, in seconds) was determined by using simple linear regression:

$$MF = IMF + LS \times \text{Time} \quad (1)$$

where IMF is the intercept and represents the initial median frequency in Hz, and LS the linear slope of the regression (Hz/s).

The EMG and measured torque data were input into an EMG-assisted biomechanical model of the spine (19,36,37) to determine the loading of the spine during the endurance test. The EMG from the 10 trunk muscles was rectified, and an RMS value was generated via a 1-s moving window for each electrode location. The curves were then normalized by the maximal EMG obtained for the respective muscle from any of the MVCs during the calibration trials. The model assumes that the force developed in the  $i^{\text{th}}$  muscle is equal to the product of the normalized EMG, physiologic cross-sectional area (PCSA), length-tension relation [ $f(l_i)$ ], velocity-tension relation [ $g(v_i)$ ], and maximum stress-generating capacity of the muscle (Gain):

$$\text{Force}_i(t) = \text{Norm. EMG}_i(t) \times \text{PCSA}_i \times f(l_i) \times g(v_i) \times \text{Gain}(t) \quad (2)$$

Because the test was isometric, the PCSA, length-tension, and velocity-tension relations were all constant. At this point, the maximal stress-generating capacity of the muscle is still undetermined. To satisfy the equilibrium

constraint, the Gain was allowed to vary as a function of time, so that the external torque about the lumbosacral disc was balanced by the internal moment generated by the 10 muscles:

$$\text{Gain}(t) = \frac{\text{External Torque}(t)}{\sum_{i=1}^{10} r_i \times [\text{Norm.EMG}_i(t) \cdot \text{PCSA}_i \cdot f(l_i) \cdot g(v_i)]} \quad (3)$$

where  $r_i$  is the moment arm from the center of the lumbosacral disc to the centroid of the  $i^{\text{th}}$  muscle. The muscle forces (Eq. 2) are then used to calculate the lumbosacral compression, anterior shear, and lateral shear forces. Because Gain is the only undetermined parameter, the effect of fatigue on the tension-generating capability of the muscle can be studied by examining this parameter.

A measure of the coactivity of antagonistic muscle groups was developed, such that:

$$\text{COAC}(t) = \frac{\sum_j \text{Flexor Torque}_j(t) + \sum_i \text{Extensor Torque}_i(t)}{\sum_i \text{Extensor Torque}_i(t)} \quad (4)$$

where  $j$  = rectus abdomini and external obliques and  $i$  = erector spinae, latissimus dorsi, and internal obliques. This parameter measured how much torque was generated in addition to what could be expected based on only extensor muscle-generated torque. Therefore if there is no coactivation of the flexor muscles, then  $\text{COAC} = 1$ . If there is as much activation of the flexors as the extensors, then  $\text{COAC} = 2$ . The RMS of the Gain, compression force, shear forces, and coactivity index was determined at each second of the test.

The value of each variable (Table 1) was calculated at 0, 50, and 100% of the endurance time. It was verified that the sampled times did not occur at a transition point. These values were tested for the effect of fatigue by using analysis of covariance (ANCOVA) with repeated measures, by using the torque as covariate (56). Post hoc analyses were conducted on those dependent variables that were shown to be significant by using a least-squares means procedure that accounted for the effect of the covariate (56).

RESULTS

The mean endurance time for the test was  $38 \pm 27$  s. The randomization of the torque levels within and between subjects was confirmed by ANOVA, which demonstrated that there was no systematic increase or decrease in the reference torque level as the test progressed (mean, 76% MVC). The ANCOVA for the effect of fatigue on the RMS relative error ( $\text{ERR}_{\text{rms}}$ ) between the reference torque and the measured torque indicated significant increases at 100% of the endurance time compared with that at the start of the test (Table 2). On the other hand, there was no change in the variability at the three times measured, as shown by the standard deviation of the error ( $\text{ERR}_{\text{std}}$ ) at 0, 50, and 100% of the endurance time.

Table 3 indicates that there was a significant effect of fatigue on the RMS EMG of the internal oblique and latissimus dorsi and on the median frequency of the right erector spinae group. Further post hoc analysis demonstrated a 25% elevation in the internal oblique and a 70% increase in the latissimus dorsi RMS EMG at the end relative to the beginning (Fig. 3). Meanwhile, only the median frequency

TABLE 1. Dependent variables used in the data analysis

Dependent variable	Definition
Task performance	
$\text{ERR}_{\text{rms}}$	RMS of relative error between reference and measured torques
$\text{ERR}_{\text{std}}$	Standard deviation of relative error
EMG analysis	
$\text{ERS}_{\text{rms}}$	RMS of the surface EMG of the erector spinae muscle—average of right and left side
$\text{IOB}_{\text{rms}}$	RMS of the surface EMG of the right internal oblique muscle—average of right and left side
$\text{LAT}_{\text{rms}}$	RMS of the surface EMG of the right latissimus dorsi muscle—average of right and left side
$\text{RERS}_{\text{mf}}$	Median frequency of the right erector spinae muscle
$\text{RIOB}_{\text{mf}}$	Median frequency of the right internal oblique muscle
$\text{RLAT}_{\text{mf}}$	Median frequency of the right latissimus dorsi muscle
Spinal loading	
GAIN	Estimated maximum muscle stress
COMP	Predicted compression force at lumbosacral joint
ASHEAR	Predicted anterior shear force at lumbosacral joint
LSHEAR	Predicted lateral shear force at lumbosacral joint
COAC	Predicted coactivity of flexor and extensor torques

RMS, root-mean-square; EMG, electromyogram.

**TABLE 2.** Results from the repeated-measures analysis of covariance (ANCOVA) test for the effect of fatigue on the quality of performance measures from the endurance test

Variable	Effect of fatigue (p value)	Endurance time [%: Mean (SD)]		
		0	50	100
ERR <sub>rms</sub> (%MVC)	<0.005	4.7 (1.7)	6.2 (1.5)	7.6 (1.5)
ERR <sub>std</sub> (%MVC) <sup>a</sup>	NS	1.7 (0.5)	2.3 (1.5)	2.3 (0.8)

ERR<sub>rms</sub>, RMS of relative error between reference and measured torques; ERR<sub>std</sub>, standard deviation of relative error; MVC, maximal voluntary contraction; NS, not significant.

<sup>a</sup>The effect of the covariate (torque) was significant for this ANCOVA.

of the erector spinae EMG decreased during the last two sampling periods.

The significant 17% decrease in the estimate of Gain at the end of the test compared with the start of the test suggested that the maximal stress-generating capability of muscle was affected by fatigue (Table 4). Likewise, the estimates of spinal loading determined from the EMG-assisted model significantly changed, despite a relatively constant torque output. For example, the post hoc tests revealed that the predicted compression force was significantly elevated at 50% of the endurance time, the final predicted anterior shear force was lower than the initial, and the final lateral shear force was greater at the end compared with that at the beginning (Fig. 4). There was virtually no change in the measured coactivity, however.

## DISCUSSION

The discovery of increased relative error as the fatigue test progressed indicated a significant decrease in motor performance. The reduction in the force-generating capability and the decline in the speed of contraction of the muscle may produce greater loading of the passive tissues of the spine in the event of an unexpected perturbation in load or position (6,54,58). There has also been

evidence that subjects will opt for alternatives that are easier, yet have less a chance of succeeding in terms of attaining a goal, when fatigued (59). Despite the decline in accuracy, subjects did not take a longer time to respond to the change in torque as they became fatigued (see limitations). The results of this investigation may influence the specification of work/rest cycles, because they show that the muscular demand may cause a decline in performance, which may ultimately increase the potential for injury.

In addition to the examination of the quality of neuromuscular performance, the experiment allowed a unique look at the muscle recruitment and spinal loading during fatigue of the trunk musculature. For instance, the internal oblique and latissimus dorsi activation increased with time, suggesting that fatigue caused greater recruitment of secondary extensor muscles to compensate for the decline in force-generating capability of the primary extensor muscles (14,29,62). The substantial increases probably cannot be explained by greater low-frequency signal passage (12) because the median frequency did not decrease significantly for these muscles.

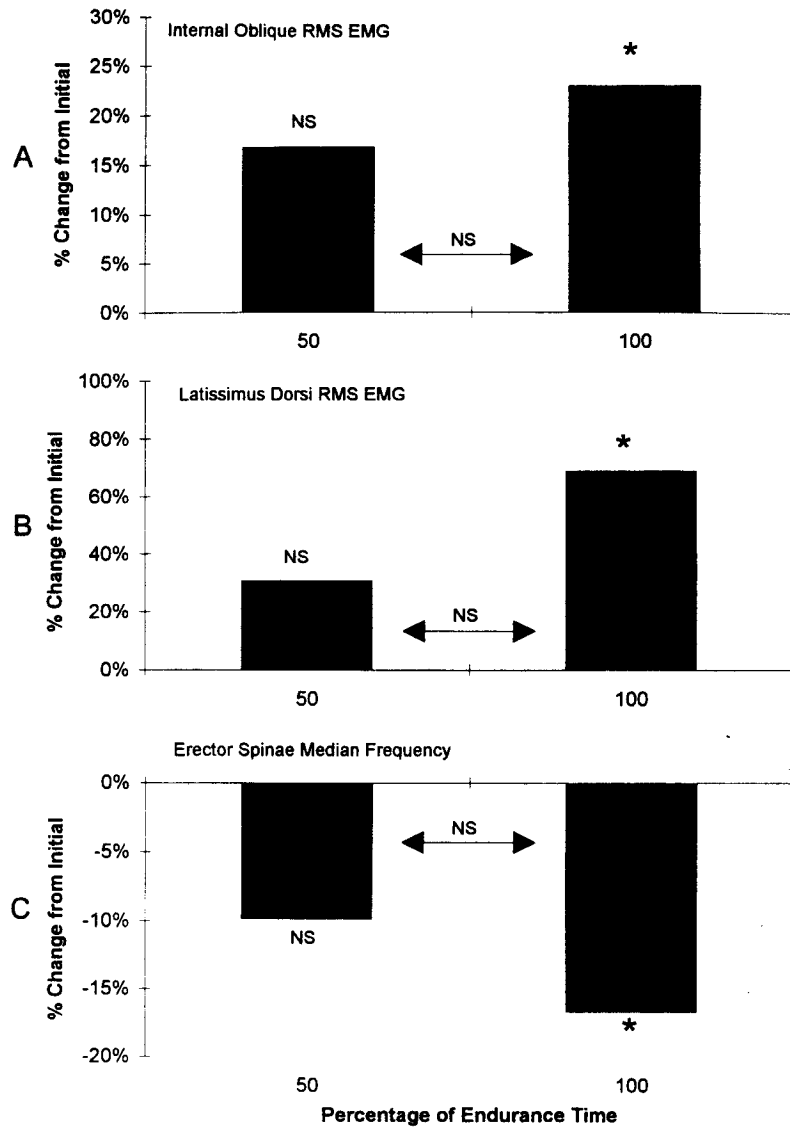
Despite the increases in internal oblique and latissimus dorsi EMGs, there was no greater activation in the erector spinae EMG in this study at a mean exertion of 75% MVC.

**TABLE 3.** Results from the repeated-measures analysis of covariance (ANCOVA) test for the effect of fatigue on the electromyographic parameters from the endurance test

Variable	Effect of fatigue (p value)	Endurance time [%: Mean (SD)]		
		0	50	100
ERS <sub>rms</sub> (%MVC)	NS	88 (10)	86 (10)	91 (10)
IOB <sub>rms</sub> (%MVC) <sup>a</sup>	<0.05	51 (15)	60 (17)	63 (16)
LAT <sub>rms</sub> (%MVC)	<0.05	31 (21)	41 (29)	53 (32)
RERS <sub>mf</sub> (Hz)	<0.005	71 (12)	64 (9)	59 (11)
RIOB <sub>mf</sub> (Hz)	NS	65 (9)	60 (16)	60 (15)
RLAT <sub>mf</sub> (Hz) <sup>a</sup>	NS	64 (18)	67 (14)	60 (19)

ERS<sub>rms</sub>, RMS of the surface EMG of the erector spinae muscle—average of right and left side; IOB<sub>rms</sub>, RMS of the surface EMG of the right internal oblique muscle—average of right and left side; LAT<sub>rms</sub>, RMS of the surface EMG of the right latissimus dorsi muscle—average of right and left side; RERS<sub>mf</sub>, median frequency of the right erector spinae muscle; RIOB<sub>mf</sub>, median frequency of the right internal oblique muscle; RLAT<sub>mf</sub>, median frequency of the right latissimus dorsi muscle; MVC, maximal voluntary contraction; NS, not significant.

<sup>a</sup>The effect of the covariate (torque) was significant for this ANCOVA.



**FIG. 3.** Percentage change in the electromyographic (EMG) measures at 50 and 100% of the endurance time compared with the initial value. Difference (\*, significant at  $\alpha = 0.05$ ; NS, not significant) from initial value indicated above bar. Difference between values at 50 and 100% indicated between bars. **A:** Internal oblique root-mean-square (RMS) EMG. **B:** Latissimus dorsi RMS EMG. **C:** Erector spinae MF.

The EMG of the erector spinae has been found to increase in some instances (53,70) and decrease in others (9,58) in response to fatigue during isometric exertions at varying levels of contraction strength.

There was a significant decrease in erector spinae median frequency at the end of the test. The decline in median frequency has been attributed to changes in conduction velocity that are a result of lower pH and muscle fiber type (12). The significant decline in erector spinae EMG power at higher frequencies confirmed the results obtained during constant-torque exertions (15,55,58) and indicated that median frequency can be used to document fatigue during varying torque isometric tests. However, caution must be exercised because multiple linear regression indicated that four of the 10 subjects did not demonstrate a significant linear decline in median frequency with

time, after accounting for torque. It is possible that in those four subjects, the variances attributed to torque and time were highly correlated. Because torque was entered into the model first, it may have explained some of the variance between median frequency and time. The insignificant median frequency changes of the other extensor muscles indicate that either they did not become fatigued sufficiently in this test or the relation between median frequency, torque, and time for these muscles is more complex. This result warns against indiscriminate use of MPF during industrial ergonomics evaluation without considering the relative load of the muscle, as suggested by others (12,58).

The results showed that Gain decreased as the subjects became fatigued. Symbolically, this means that for a given level of torque, there was greater muscle activation at the

**TABLE 4.** Results from the repeated-measures analysis of covariance (ANCOVA) test for the effect of fatigue on the measures determined from the EMG-assisted model of spinal loading

Variable	Effect of fatigue (p value)	Endurance time [%: Mean (SD)]		
		0	50	100
GAIN (N/cm <sup>2</sup> ) <sup>a</sup>	<0.05	52 (16)	49 (16)	43 (13)
COMP (N) <sup>a</sup>	<0.05	1949 (361)	2026 (397)	1969 (383)
ASHEAR (N) <sup>a</sup>	<0.005	799 (271)	740 (264)	681 (267)
LSHEAR (N)	<0.05	80 (59)	120 (108)	142 (105)
COAC	NS	1.14 (0.08)	1.15 (0.08)	1.13 (0.08)

GAIN, estimated maximum muscle stress; COMP, predicted compression force at lumbosacral joint; ASHEAR, predicted anterior shear force at lumbosacral joint; LSHEAR, predicted lateral shear force at lumbosacral joint; COAC, predicted coactivity of flexor and extensor torques; EMG, electromyogram; NS, not significant.

<sup>a</sup>The effect of the covariate (torque) was significant for this ANCOVA.

end of the test. Looking at the recruitment, it can be seen that although the erector spinae activation remained the same, there was an increase in latissimus dorsi and internal oblique activity. The finding of decreased Gain can be interpreted in two ways. If we assume that the maximal stress-generating capability is constant, then we could incorporate a fatigue-modulating factor into Eq. 2 that accounts for the decline in torque output for a given level of activation. Alternatively, we could say that the maximal stress that a muscle can produce decreases with fatigue and leave Eq. 2 as it is. It is important to stress that if we had assumed a constant value of Gain, then our estimate of spinal loading would have been off by as much as 20%, given the other assumptions of the model. For more rigorous mathematic developments, theoretic methods for the inclusion of a fatigue-modulating factor have been developed by Hawkins and Hull (24) and Giat et al. (17).

The current EMG-assisted model assumed that all muscles had equal stress-generating capacity. It is not clear why muscles of different fiber-type proportions, muscular architecture, and functional anatomy must have the same maximal muscle stress. The use of a single Gain for all muscles is undoubtedly simplistic. However, because of the scarceness of data and the invasive nature of the biopsy needed to determine the effect of muscle-fiber distribution on the maximal stress capacity, the use of single Gain is justified. Thelen et al. (68) used system-identification techniques for functionally apportioning the maximal muscle stress. Lack of direct measurement of muscle forces keeps such elegant and logical mathematic assumptions untested and speculative.

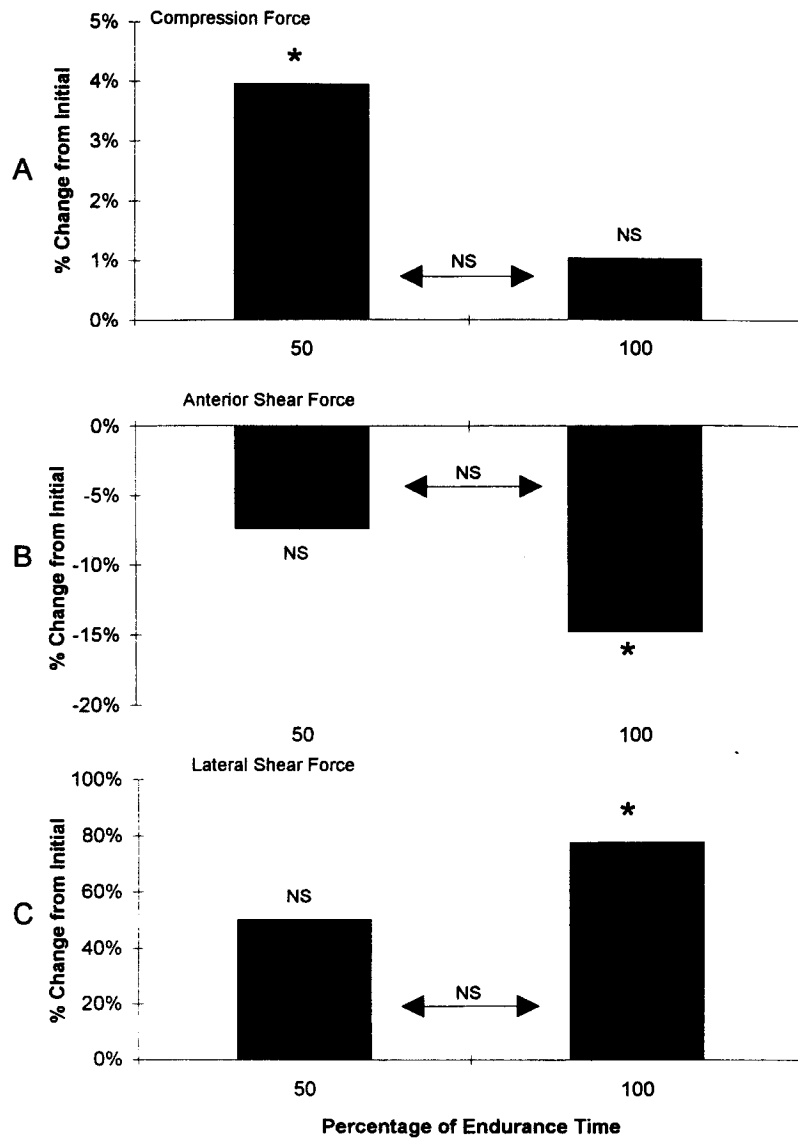
The changes in recruitment patterns result in spinal-loading gradients despite relatively constant torque output. Increased compressive loading due to fatigue could have an additive effect on the increased risk of injury. With higher loads, fewer vertebrae can withstand the loads (7). Additionally, with sustained cyclic loading, there is

decreased tolerance of the vertebral endplates caused by fatigue (7,30). The presence of an increasing lateral shear during fatigue signifies asymmetry in bilateral muscle recruitment. Asymmetric loading generally increases fiber strain in the intervertebral disc, possibly leading to fiber rupture, and also increases the facet-joint contact forces (60,61). On the other hand, the changes in recruitment may also provide protection against excessive passive tissue loading by increasing the muscular endurance through alternating activity (14).

During the tests, there were periods of reduced EMG activity for five of the 10 subjects. Figure 5 shows the EMG activities for a 5-s period taken from between seconds 4 and 9 of the test depicted in Fig. 2. Figure 5A shows a reduction in the force generation. Furthermore, in Fig. 5B and C, there is a substantial reduction in the raw erector spinae EMG in the area between the arrows. The reduced muscle activity may have been an attempt to increase the synchronization of the motor units (44,45) so that the high torque demand could be maintained.

The reduced muscle activity also could be related to pre-movement silence periods that have been shown to precede rapid ballistic movements in limb muscles (2,27). It is uncertain whether this is the same effect, because we do not believe that it has been shown before for the erector spinae during an isometric contraction. We also could not verify whether the muscle became completely silent. This type of event occurred also during the submaximal training sessions, so it is not likely that this is purely a fatigue-mediated response. Nonetheless, this period of reduced muscle activity at a time of high torque demand may leave the spine unprotected and vulnerable to injury. Future research may elucidate this occurrence more fully. This inherent variability of the muscle activation demonstrates the value of EMG-assisted models; optimization-based models would not have been able to predict this potentially significant event.





**FIG. 4.** Percentage change in the spinal loading measures at 50 and 100% of the endurance time compared with the initial value. Difference (\*, significant at  $\alpha = 0.05$ ; NS, not significant) from initial value indicated above bar. Difference between values at 50 and 100% indicated between bars. **A:** Predicted compression. **B:** Predicted anterior shear. **C:** Predicted lateral shear force.

**LIMITATIONS AND FUTURE RESEARCH**

This study represents the first step in trying to understand the effect of fatigue on the performance of the trunk musculature. As such, the format of the test is undergoing minor modifications. We believe that the format of the test, as presented here, stressed the muscular system too greatly. As a result, subjects became fatigued in a very short time. This prevented the collection of a sufficient number of data points for the determination of the response time as a function of fatigue for each subject. Consequently our hypothesis, that the response of the muscles to a change in torque would be reduced, was rejected. However, we believe that modifications in the testing procedure will allow us better to assess this proposition.

At this point, the results from isometric endurance tests cannot be generalized to tasks that involve dynamic trunk extension. It is recognized that trunk strength is task specific (39,48). There is no reason to believe trunk endurance is not the same. Consequently, further investigation of the trunk muscle-recruitment patterns during dynamic endurance tests is warranted.

The estimation of the spinal loading has been shown to be sensitive to many factors. For example, the geometry of the muscles may supply a large source of variation (47,50). The difficulty in obtaining a true MVC and the variability in EMG when obtaining an MVC also must be considered (16,41). It is also certain that muscles of different fiber-type proportions will certainly become fatigued at different rates. Therefore the use of a single Gain for all muscles is undoubtedly simplistic. However, because of

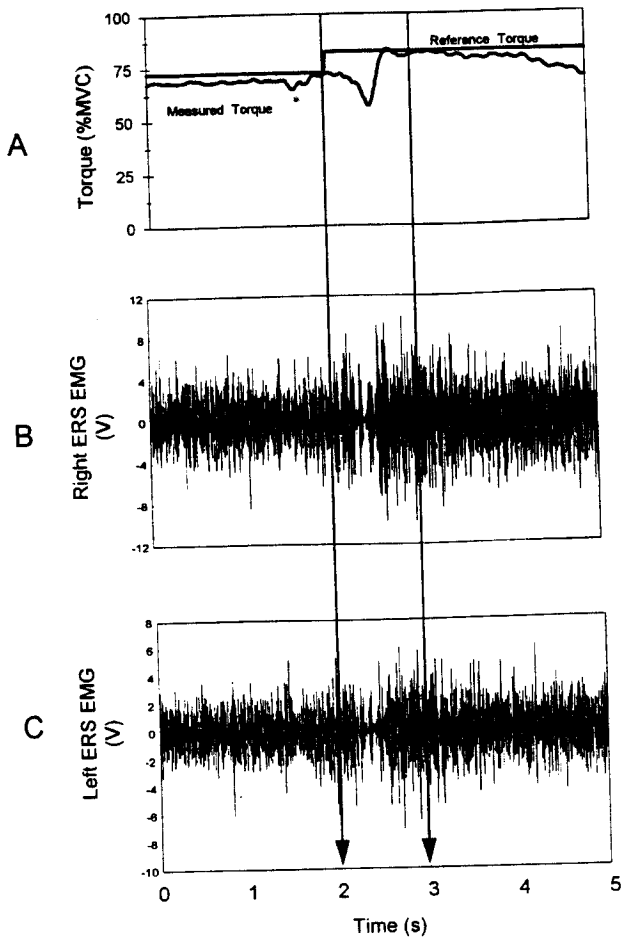


FIG. 5. Reduced electromyographic (EMG) period. A: Measured and reference torque recording. B: Raw right erector spinae EMG. C: Raw left erector spinae EMG.

the scarcity of data and the invasive nature of the biopsy needed to determine the subjects' fiber distribution, the maximal stress capacity for all muscles was assumed to be equal. Thelen et al. (68) used parameter estimation techniques for functionally apportioning the maximal muscle stress.

In addition, all the muscles were assumed to have linear EMG-tension relations. The linear EMG-tension relation has been well documented for the erector spinae (1,9). There also has been discovery of a nonlinear EMG-tension relation in recent years (52,65,67). It is important to stress that the nonlinearity may be the result of the recruitment strategy within the multiple-muscle system and not necessarily attributable to the EMG-tension relation of the muscle. Woods and Bigland-Ritchie (73) demonstrated that the EMG-tension relation may be related to muscle-fiber type. Moreover, nonlinear EMG-tension relations for the other muscles present in the model may have

occurred. To obtain better estimates of spinal loading, more robust determination of the EMG-tension relation is warranted. However, this would have entailed many submaximal calibration contractions for each muscle, which might have induced a level of fatigue before the main test.

Finally, it is important to recognize that the findings about the response of Gain apply to only these long-duration isometric exertions. The EMG-assisted model was used primarily to infer the changes in spinal loading due to a fatiguing task. Application of the model to conditions for which the model was intended—short-duration dynamic lifting exertions—results in greater stability of the Gain (19). However, there is some evidence that, for an extended session of repetitive lifting, the Gain may require adjustment over the course of the session because of the effects of muscle fatigue (20).

## CONCLUSIONS

A novel testing protocol was used to investigate changes in neuromuscular performance, muscle recruitment, and spinal loading as subjects became fatigued while performing an isometric endurance test of varying torque requirements. There was decreased accuracy in maintenance of a reference torque and no change in response time as subjects became fatigued. The study of trunk-muscle recruitment indicated significant increases in internal oblique and latissimus dorsi muscle activity. This change in recruitment led to gradients in spinal loading, despite a relatively constant torque output. The use of an EMG-assisted model demonstrated that when subjects are expected to become fatigued during test performance, the assumption of a constant maximal stress capacity of the muscle may not be robust. The observation of a reduction in muscle activity when the torque demand is quite high may signify a potential injury mechanism and should be further studied.

**Acknowledgment:** We thank the Biodynamics Lab staff for their invaluable assistance and support from NIDRR, RERC grant H133E0009, Ohio Bureau of Worker's Compensation, Division of Safety and Hygiene, and the Davis Medical Research Grant.

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