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Trunk Muscle Fatigue and Its Implications in EMG-Assisted Biomechanical Modeling

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Abstract

Muscle fatigue affects the underlying EMG-force relationship on which EMG-assisted biomechanical models rely. The aim of this study was to evaluate the impact of short duration muscle fatigue on the muscle gain value. Participants performed controlled, isometric trunk extension exertions at 10, 20, and 30 degrees of trunk flexion and controlled isokinetic trunk extension exertions at 5 and 15 degrees/sec on five separate days. Fatigue of the lumbar extensors was generated by moderate-intensity, trunk extension exertions. Participants performed controlled test contractions at defined intervals throughout the fatiguing bout and the EMG activities of trunk muscles were collected. These EMG data were employed in a standard EMG-assisted model and fatigue-dependent trends in the gain factor were evaluated. The results of this study show a significant effect of fatigue on the gain factor. Further analysis revealed that by reducing the magnitude of the gain factor in proportion to the drop in median frequency of the EMG signal of the fatigued muscle, a more accurate estimate of muscle force can be calculated. Mean normalized error between predicted internal moments and the measured external moments improved from 17.5 to 9.6% error via the implementation of this modified gain factor.

Keywords: EMG-assisted Model, Fatigue, Gain Factor, Lumbar Extensors

Relevance to Industry: Biomechanical models are used to quantify low back stress during occupational lifting tasks. Understanding how these models need to be modified under conditions of muscular fatigue will provide more accurate estimates of stress and can help reduce the incidence of low back injury.

Highlights:

- Muscular fatigue affects predicted tissue stress of EMG-assisted models
- Gain value of EMG-assisted models needs to be modified as a function of fatigue
- Change to gain factor is linearly related to the change in the EMG median frequency

INTRODUCTION

Over the last thirty years, the EMG-assisted biomechanical modeling technique has been developed and refined for the lumbar spine (e.g. McGill and Norman 1986; Marras and Granata 1997; Davis and Mirka 2000; Gagnon et al., 2011). The goal of these models is to use empirical muscle activation data captured through the use of electromyography (EMG) to estimate the force produced by the trunk musculature and thereby develop an accurate estimate of the spine reaction forces. The basic model utilizes the EMG activity of the muscle (expressed in terms of % of the muscle's capacity, NEMG), the physiological cross-sectional area (PCSA) of the muscle, a maximum muscle stress value (expressed in newtons per unit of cross-sectional area and herein refereed to as the "gain" factor), and modulation factors based on the instantaneous muscle length (the force-length (F-L) modulation factor) and the instantaneous velocity of shortening of the muscle fibers (the force-velocity (F-V) modulation factor) in a multiplicative model to calculate an estimate of the force transmitted through the tendon of the muscle to their bony attachment sites. This value can then be multiplied by the appropriate moment arm (MA) to generate an estimate of the moment contribution of this muscle to the net internal moment:

$$\text{Muscle Moment (Nm)} = [\text{NEMG}] * [\text{PCSA}] * [\text{Gain}] * [\text{F-L}] * [\text{F-V}] * [\text{MA}] \qquad (1)$$

An underlying assumption of these EMG-assisted biomechanical models is that the relationship between this electromyographic activity and the muscle force is linear and stationary – an assumption that is violated as the muscles experience fatigue. It has been shown for multiple muscles and a variety of conditions that, as skeletal muscle fatigues, the electrical activity of the muscle increases while maintaining a constant force. (See Luttmann (1996) for a discussion of the underlying physiological processes that generate these changes.) These alterations in this relationship will impact the gain factor in EMG-assisted biomechanical models. Sparto and Parnianpour (1998) attempted to address this challenge to the EMG-assisted modeling technique by employing frequent maximum voluntary contraction (MVC) exertions to re-calibrate the EMG-force relationship to account for this fatigue "artifact" in the EMG signal. Their results showed that this repetitive recalibration of the EMG data provided the necessary alteration to the EMG-force relationship to establish reliable estimates of the muscle forces during the fatiguing exertions. Of course, the major methodological difficulty of multiple MVC exertions interspersed throughout the task are two fold: these multiple MVC exertions could be potentially risky for participants and the simple act of performing these MVC exertions may artificially accelerate the rate of muscular fatigue development. A new methodology for recalibration of this EMG-force relationship under fatiguing conditions would allow for studies that can provide deeper insights into spine loading during times of significant muscle fatigue.

The objective of this study was to explore the relationship between muscular fatigue and the gain factor. If this factor is found to vary as a function of fatigue level, the second objective of this research was to develop a method for modifying the factor in a way that would allow the model to predict the muscle forces in both the unfatigued and fatigued states.

METHODS AND MATERIALS

Participants

Six male participants were recruited from a PhD candidate population. None had chronic musculoskeletal problems of the low back or lower extremity and none were currently experiencing discomfort in these regions. The basic anthropometry of the participants (mean (± SD)), was: age 29 (± 3.3) years, height 176.5 (± 3.7) cm, whole body mass 71.5 (± 4.5) kg, trunk depth 21.8 (±1.7) cm, and trunk width 31.0 (1.8) cm. All provided written informed consent before participation (approved by the Institutional Review Board of Iowa State University).

Apparatus

Electromyographic activity of trunk muscles was recorded using bipolar surface electrodes (Model DE-2.1 Bagnoli™ from DelSys, Boston, MA). These EMG data were collected bilaterally from rectus abdominis, external obliques, erector spinae and latissimus dorsi muscles. The sampling locations were as follows: rectus abdominis - 3 cm above the umbilicus and 3 cm lateral to midline; erector spinae - 4 cm lateral to the L3 spinous process; external oblique - 15 cm lateral to the umbilicus and in the midpoint between the 12th rib and iliac crest; latissimus dorsi – T8-T12 level over the belly of the muscle as established though palpation.

A Kin/Com lumbar dynamometer (Chattanooga Group, Inc., Hixson, TN, USA) and an aluminum reference frame were used to create a trunk extension exercise environment that allowed for the precise control of trunk angle, trunk extension velocity and trunk extension moment (Mirka and Marras, 1993). The angle, velocity, and moment of the trunk flexion/extension data were collected from Kin/Com and were synchronized with the EMG signal at 1024 Hz by using the "MotionMonitor" data acquisition software (MotionMonitor, Innovative Sports Training, Chicago, IL, USA). The participants were required to control their level of the trunk extension moment during some of the experimental trials using a video feedback system that displayed a target moment and the instantaneous moment being generated by the participant. In this way the participants were able to maintain the designated amount of torque to within +/- 10% of the target moment. If they were unable to meet this requirement, the condition was repeated.

Independent variables

There were two independent variables for isometric trials: trunk flexion angle at three levels (ANGLE: 10°, 20°, and 30°), and time into the fatiguing trunk extension exertion protocol (TIME: 0 (pre-fatigue), 15, 30, 45, 60, 75 s). The isokinetic trials had two independent variables as well: trunk angular extension velocity at three levels (VELOCITY: 0, 5 and 15°/second) and TIME: 0, 15, 30, and 45 s. To provide the data for the 0°/second condition, the data from the isometric trials were utilized (the 20° flexion condition). While data were collected up to 75 seconds for the isometric trial where possible, not all participants were able to maintain the exertion for that long, so subsequent data analysis will only consider up to 60 seconds.

Dependent variables

The dependent variables were the normalized EMG (NEMG) of sampled bilateral muscles: erector spinae, latissimus dorsi, rectus abdominis and external oblique. These NEMG activities of the trunk muscles were used as inputs for an EMG-assisted biomechanical model for the prediction of muscle forces, net internal moment, and, ultimately, the spinal loads.

Experimental procedure

In the first session, anthropometric data were recorded and the participant performed a brief warm up. The participant then learned how to use the video feedback system to maintain the required trunk extension moment during the isometric and isokinetic experimental trials. Upon completion of the training, participants were fitted with surface electrodes as described in the Apparatus section. The participant then stepped into the asymmetric reference frame dynamometer and their pelvis was secured to the dynamometer using a belt harness. They then performed one of the five experimental conditions. The order of presentation of the conditions was randomized across days.

On the three days when an isometric condition was tested, the participant performed two trunk extension and flexion maximum voluntary contractions (MVCs) against the static resistance provided by the dynamometer in the trunk posture for that condition (ANGLE set at either 10, 20, or 30 degrees of trunk flexion from a upright standing posture as specified for that condition). EMG signals were recorded for 3 seconds during the MVC trial. A two-minutes rest period was provided after each maximum voluntary exertion. In the subsequent experimental trials, the participant generated a continuous 50% MVC isometric extension force against resistance provided by the dynamometer, and controlled through the visual feedback display (error of less than +/- 10% tolerance was required). EMG data were collected at time 0 seconds and then recorded every 15 seconds. The duration of EMG data collection was 4 seconds (0-4s, 15-19s, etc.). The participant maintained this exertion until exhaustion.

On the two days that an isokinetic condition (5 or 15°/second) was tested, the same preliminary (warm-up, electrode placement, etc.) were followed and MVC exertions were performed in the 20° trunk flexion angle. In the subsequent experimental trials, the participant generated a continuous isometric extension force equal to 50% MVC for 15 seconds at 30° trunk flexion, followed by performing a dynamic 50% MVC exertion while moving at the designated velocity from 30 to 10 degrees of trunk flexion. The required trunk extension moment during these isokinetic extensions were again controlled through the visual feedback system. The angular velocity was maintained by the dynamometer. After completing this test trial the participant then returned to the fatiguing isometric exertion for another 15 seconds followed by another 50% MVC dynamic exertion. This protocol was repeated until exhaustion with the EMG data being recorded during the isokinetic concentric contraction and three seconds of the isometric exertion (isometric data used to assess muscle fatigue using median frequency methods).

Data processing

Unprocessed EMG data from all sampled muscles was first transformed into the frequency domain and then band-pass filtered (10 – 500 Hz) and notch filtered at 60 Hz (and the aliases up to 500 Hz). The median frequency of the EMG data of the erector spinae for all isometric exertions was then computed to monitor fatigue development. The data were then transformed back into the time domain, rectified (full-wave) and normalized (to maximum). For the isometric exertions, the data of interest were those collected during the 4-seconds data collection periods. For the isokinetic exertion, the data of interest were the EMG collected as the participant passed through a trunk flexion window of 22 to 18 degrees. Since these exertions were all sagittally symmetric, the mean of the two bilateral muscles was computed and employed in the subsequent analyses.

Analysis to explore fatigue effects on gain factor

To explore the effect of short duration fatigue on gain factor, it was not sufficient to simply demonstrate a significant effect of TIME on the magnitude of the NEMG of the erector spinae. If there were a significant increase in the NEMG as a function of fatigue this could simply be in response to an increase in the level of antagonistic muscle forces from the rectus abdominis and the external obliques. Therefore it was necessary to utilize an EMG-assisted biomechanical model and assess the quality of the predictions of the model with and without fatigue-dependent changes in the gain factor for the erector spinae muscles while keeping all other gain factors constant. The EMG-assisted model developed by

Marras and Granata (1997) was used for this purpose. First, for each participant, we developed an EMG-assisted model reflecting the pre-fatigue condition and found the gain value that created equilibrium between the internal and external moments and we call this $Gain_{PRE\text{-}FATIGUE}$. For all subsequent points in time this gain value was used for the latissimus dorsi, rectus abdominis, and external oblique for that participant. The gain factor for the erector spinae was allowed to change to maintain equilibrium between internal and external moments in all subsequent data collections in the trial. In addition to NEMG data, the median frequency of the erector spinae was tracked as a function of time. A least squares analysis was used to develop an equation that would predict the gain value to be used under conditions of fatigue utilizing this median frequency (MF) value:

$$Gain_{PREDICTED} = (-0.0592 + 1.046\ MF_{CURRENT}/MF_{PRE\text{-}FATIGUE}) * Gain_{PRE\text{-}FATIGUE} \quad (2)$$

Where:

$Gain_{PREDICTED}$ is the predicted gain in the fatigued state
$Gain_{PRE\text{-}FATIGUE}$ is the gain observed in the unfatigued state (Time = 0)
$MF_{CURRENT}$ is the median frequency of the muscle in its fatigued state
$MF_{PRE\text{-}FATIGUE}$ is the median frequency of the muscle in the unfatigued state (Time = 0)

The value of $Gain_{PREDICTED}$ was compared with a value of $Gain_{VARIANT}$ (the trial-specific gain required to produce equilibrium between internal and external moments). The Pearson correlation coefficient describing the relationship between $Gain_{PREDICTED}$ and $Gain_{VARIANT}$ across all conditions and across all participants was 0.79. Using the values of $Gain_{PREDICTED}$, the net internal moments and spine reaction forces were calculated, as were those calculated using the traditional $Gain_{PRE\text{-}FATIGUE}$ value. We then computed the mean absolute error (MAE in Nm) and mean normalized error (MNE expressed in terms of percent error) of the net internal moments using both variant and invariant methods.

$$MNE=100*[\text{Estimated internal moment- External moment}]/\text{External moment} \quad (3)$$

Model validation

The purpose of model validation was to explore the effectiveness of the approach shown in Equation 2, to predict muscle forces for different participants under a combination of old and new fatiguing trials. Two new male participants (mean (± SD), was age 31.5 (± 3.54) years, height 179 (±

8.49) cm, and whole body mass 75.5 (± 3.54) kg) were recruited and they performed the isometric extension trials at 20 and 25 degrees of trunk flexion and the isokinetic trials at 5 and 10 degrees/sec. All other procedures were those described in the Methods and Materials.

RESULTS

Figure 1 demonstrates that the goal of introducing fatigue in the lumbar erector spinae was achieved by showing a consistent drop in the median frequency of the EMG. The ANOVA revealed that the median frequency of the erector spinae was significantly ($p<0.05$) reduced by TIME. Figure 1 also shows the results of our study confirm a negative shift of the median frequency with increased muscle elongation as a result of increased trunk flexion angle (Mannion and Dolan, 1996). The lower levels of median frequency shown in the 5 deg/sec level in Figure 2 was unexpected but was a consistent response across participants.

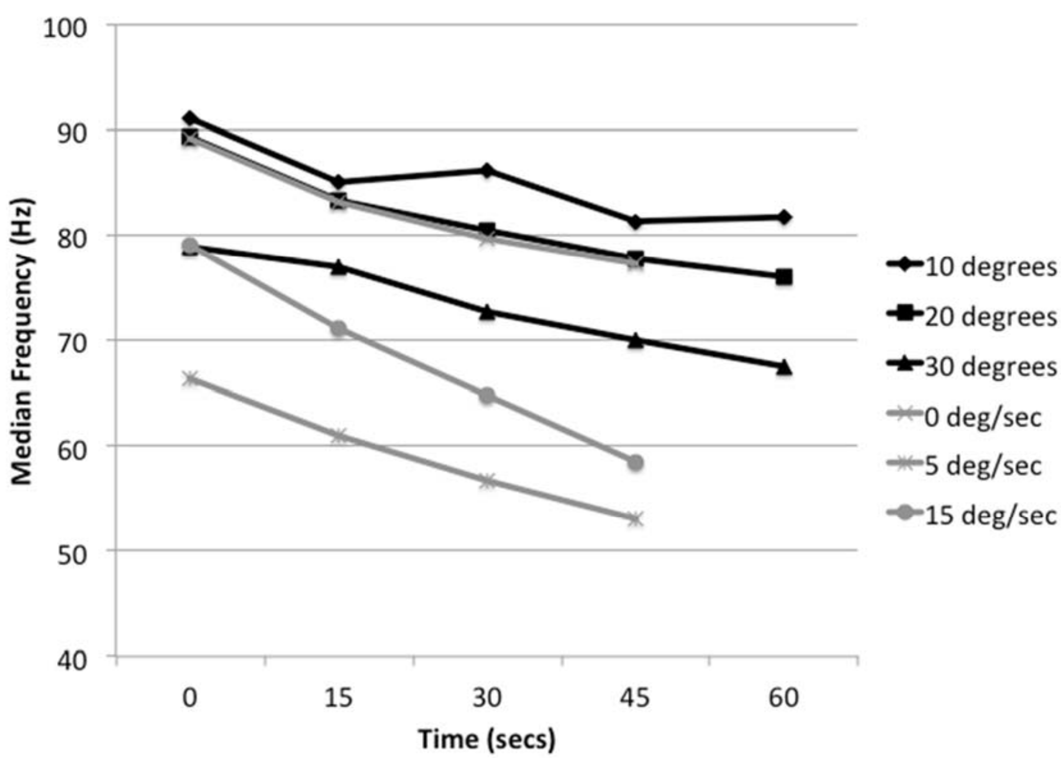

Figure 1. Erector spinae median frequency as a function of TIME under different posture and motion conditions.

To explore the impact of allowing the gain factor to vary as a function of the change in median frequency, a comparison was made between the standard EMG-assisted model using a single pre-fatigue gain value and the modified model that allows the gain to vary as a function of fatigue (Equation 2). The results showed that the mean absolute error was reduced from 18.6 Nm to 10.2 Nm when the gain was adjusted based on median frequency shift, resulting in a mean normalized error reduction from 17.5% to 9.7%.

Finally, the impact of these differences on the net computed moments and estimated spine reaction forces are shown in Figures 2-4 for the isometric conditions. In these figures "Actual" refers to the net internal moment that is computed if the gain value is computed for each trial (using $Gain_{VARIANT}$) and would be considered the gold standard for comparison as it allows the value of the gain for the erector spinae muscle to vary to provide an exact match between the internal and external moments. "Invariant" and "Predicted" are those moment / forces predicted using the two approaches evaluated in the analysis above. This analysis of moments and forces provides an appreciation for the magnitude of the errors in these predictions that can be avoided by using this variable gain factor approach that predicts muscle force using a gain that is predicted by the drop in median frequency. It is noteworthy that the value of spine compression is not significantly affected by fatigue level when this modification to the gain factor is applied. This result is in conflict with several previous studies that have indicated muscle fatigue can increase spine compression, and highlights the importance of the gain modification supported by the results of the current study.

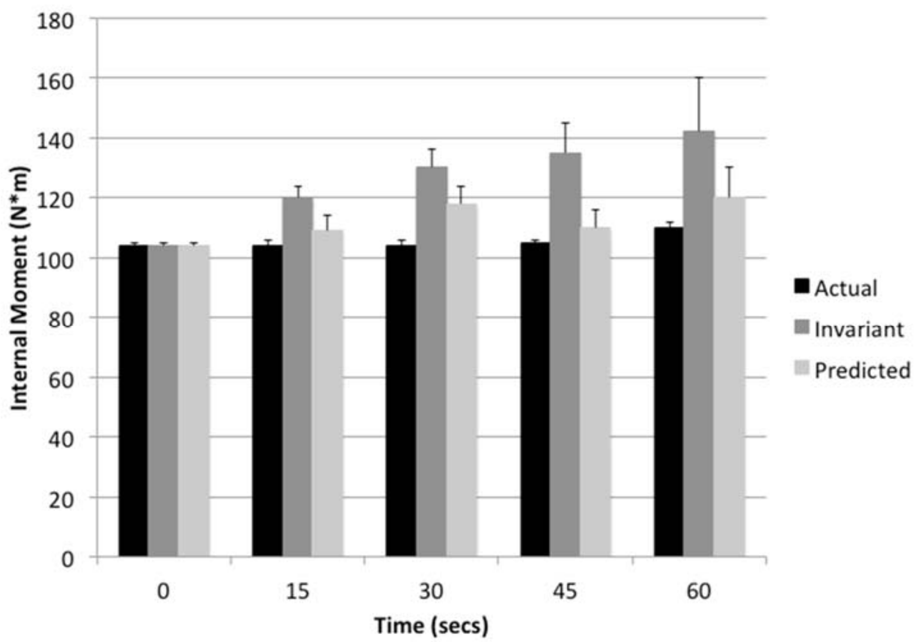

Figure 2. Calculated net internal moments about L3 using three different methods of setting the gain for the erector spinae: 1) Actual – gain set to create equilibrium between internal and external moment on a trial-by-trial basis; 2) Invariant – gain set in the unfatigued condition and used in all subsequent conditions; and 3) Predicted – gain set by using method proposed in the current study (Equation 2). Standard error bars shown.

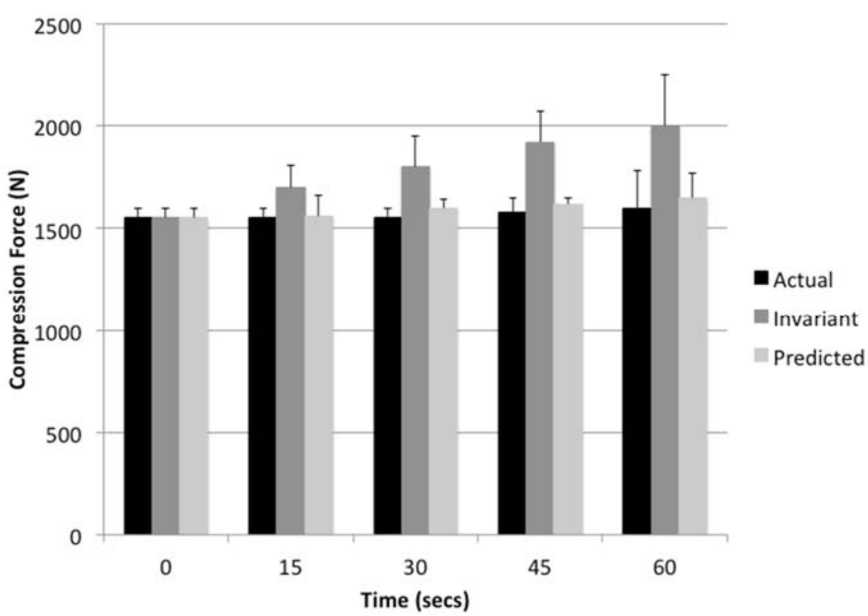

Figure 3. Calculated resultant spine compression force at L3 using three different methods of setting the gain for the erector spinae: 1) Actual – gain set to create equilibrium between internal and external moment on a trial-by-trial basis; 2) Invariant – gain set in the unfatigued condition and used in all subsequent conditions; and 3) Predicted – gain set by using method proposed in the current study (Equation 2). Standard error bars shown.

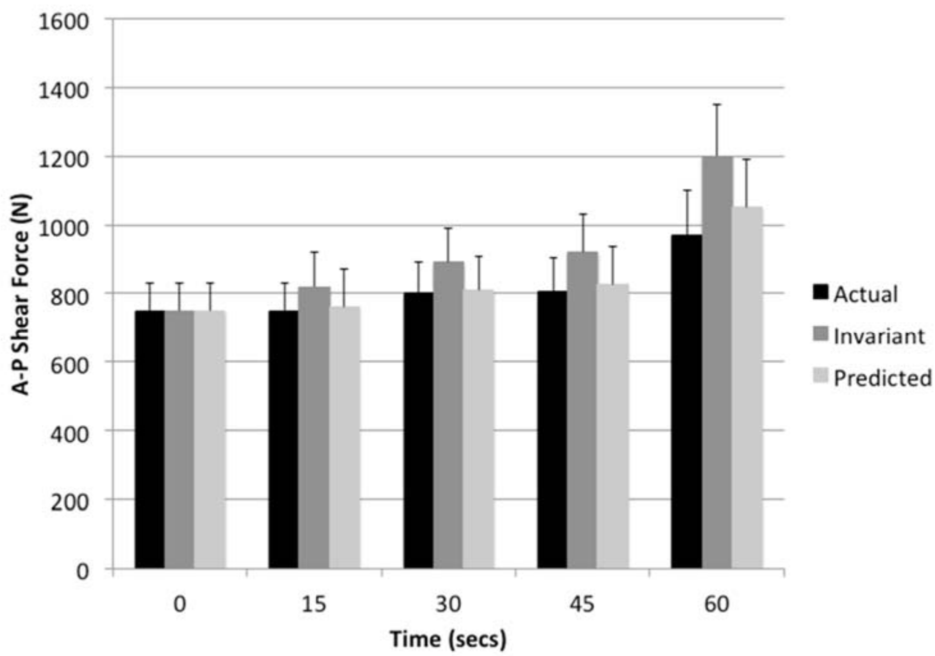

Figure 4. Calculated resultant anterior-posterior shear force at L3 using three different methods of setting the gain for the erector spinae: 1) Actual – gain set to create equilibrium between internal and external moment on a trial-by-trial basis; 2) Invariant – gain set in the unfatigued condition and used in all subsequent conditions; and 3) Predicted – gain set by using method proposed in the current study (Equation 2). Standard error bars shown.

Model validation results

The results of the analysis of the data from the two new participants and new conditions confirm a reduction in both MAE and MNE with the new method and provide reasonable levels of error in these predictions of internal moments. The results showed that the mean absolute error was reduced from 22.5 Nm to 13.3 Nm when the gain was adjusted based on median frequency shift, resulting in a mean normalized error reduction from 21.4% to 12.9%. As expected, these normalized error values are a bit higher than those seen in the original data set but do show a significant improvement in predictions for these new participants.

DISCUSSION

This study explored the effect of muscular fatigue on the gain factor of EMG-assisted biomechanical models. Our results show that, under these specific fatiguing conditions, the gain factor (force/unit area) in these EMG-assisted biomechanical models is significantly impacted by short duration muscular fatigue and therefore must be modified based on the level of fatigue in the muscles. An algorithm for making this modification has been presented and results of its use in predicting muscle forces have shown very positive results.

EMG-assisted biomechanical spine models have historically relied on a single value of gain for all contributing muscles (e.g. Granata and Marras 1993; Granata and Marras 1995; Marras and Granata 1997) and this has been shown to be very effective for most lifting scenarios. However, as muscular fatigue develops force generation capability of fatigued muscle declines and the magnitude of the EMG signal has been shown to increase at constant force levels (e.g. Sparto and Parnianpour 1998). These observations demonstrate that a constant gain factor must be reconsidered under fatiguing conditions. Sparto and Parnianpour (1998) developed a technique for this recalibration problem. They measured maximum force generation capacity frequently throughout a fatiguing exertion and then used the change in force producing capability to modulate the gain factor of the erector spinae. While we recognize this as a sound approach to meet this recalibration need, ideally we would like to have this recalibration done without the need for these frequent MVC exertions that can accelerate the development of fatigue beyond the fatiguing exertion being studied. The results of the present study support a new method to provide this recalibration through the use of median frequency change as the predictor (Equation 2) instead of change in force producing capability. Our results showed that estimation of the net internal moment with a constant (invariant) gain factor leads to an overestimation this net internal moment of more than 20%. When this gain factor is altered using the algorithm developed in our study the internal moment is reduced to more accurate levels and the improvement of the mean absolute error and normalized error return these levels of accuracy to those seen in other EMG-assisted model that do not consider the muscle fatigue. These data suggest that the model developed in this study is an effective means of resolving the gain-flux problem under certain fatiguing conditions.

The results of our study have potentially important implications when studying spine reaction forces under conditions of significant muscle fatigue. Since the spine reaction forces (compression, anterior-posterior shear, lateral shear) are the net result of the internal muscle forces, over-estimation of the force produced by fatigued muscles because of this fatigue artifact can lead to misleading conclusions. In the current study for example, the invariant gain factor approach lead to the conclusion that there was a 20% increase in compression force. When the Equation 2 algorithm was applied to the erector spinae muscle force calculation, compression force still increased (due to the increased activity of the antagonist muscles as shown by others (e.g. Potvin and O'Brien, 1998; Granata et al., 1999), but the magnitude of this increase was much less than the 20% predicated by the invariant gain approach. Interestingly, the A-P shear force increased in both the variant and invariant gain approaches, further illustrating the increase in antagonist (here external oblique) muscle forces. Collectively, these results point to the need for EMG-assisted biomechanical models to adjust muscle force estimates by modifying the gain factor as a function of fatigue. The results of this study have shown that modifying these gains based on the magnitude of the median frequency shift is an effective way to achieve this aim.

There are several limitations to the generalizability of these results that should be recognized. First, the EMG-assisted biomechanical model chosen for this study is quite simple by today's standards. While this would not limit the interpretation of our results, a more advanced EMG-assisted biomechanical model may provide the opportunity to be more precise in the development of an equation relating change in erector spinae median frequency to the necessary recalibration of the gain factor. Second, as a first attempt to model these phenomena, we have assumed a stationary and linear response over time and this is not likely to be true under all levels, or types, of muscular fatigue. In particular, the extension of these results to more realistic work scenarios is quite unclear. Fatigue developed in this study was over a period of minutes. The impact of changes in the median frequency of the EMG signal on the gain factor over longer periods is not at all clear. Exploration of multiple fatigue scenarios, particularly the impact of the duration of the fatigue development, would be particularly useful as this tool is considered for more realistic occupational scenarios.

CONCLUSIONS

This novel modification to a standard EMG-assisted modeling technique can lead to improved accuracy in estimates of spine loading during fatiguing exertions, and requires only a modification to the value of the gain factor using information about the drop in median frequency. Our results showed that an increase in co-activation force of antagonist muscles under fatigued conditions led to an increase in anterior-posterior spinal shear forces, while compression force predicted using our modified gain factors increased to a much lesser degree. This is in contrast to previous studies that have shown much larger increases in compression and this contrast highlights the value of the technique established in this research.

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