Length-tension and velocity-force relationships of the torso extensors: Dynamic biomechanical modeling considerations.

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Abstract

This study investigated the length-tension and velocity-force relations of the torso erectors. A myoelectric based approach was used wherein a dynamic biomechanical model incorporating active and passive tissue characteristics provided muscle kinematic estimates during controlled sagittal plane extension motions. A double linear optimization formulation from the literature provided muscle tension estimates. The data supported a linear length-tension relation toward full flexion for both the erector spinae and latissimus muscles. Velocity trends agreed with that predicted by Hill's exponential relation. The results have implications for muscle tension estimation in biomechanical torso modeling, and suggest a possible low back pain injury mechanism.

Introduction

Computerized biomechanical models of the torso continue to be developed for the investigation of low back pain (McGill and Norman, 1986; Jäger and Luttmann, 1989; Marras and Sommerich; 1991). These models attempt to accurately predict the forces on the lumbar spinal column as they have been associated with low back pain risk (Chaffin and Park, 1973; Marras et al. 1993). Toward this end, the accurate activity estimation of the modeled muscles is important since the tensions produced by these tissues are among the largest contributors to the spinal compression force (McGill and Norman, 1986).

Muscle length and shortening velocity have been demonstrated to affect maximum muscle tension (Hill, 1938; Komi, 1973). Therefore, these factors need to be accurately represented if a realistic estimate of muscle tension from muscle EMG is desired during dynamic tasks. However, current dynamic (and static) models vary greatly in the implementation of these modulators.

The purpose of this study was to investigate the torso extensor muscle length-tension and force-velocity relationships *in vivo*.

Based on the empirically observed increase in torso extension strength toward full flexion (Schanne, 1972; Graves et al. 1990), it was hypothesized that the length-tension curve, as measured using myoelectric to force relationships and accounting for passive tissue effects, increases monotonically toward full flexion. Second, although no specific evidence of concentric velocity-tension mis-modeling has been reported, this modulator was investigated to provide data with which to compare the currently used techniques.

Methods

An anatomically detailed biomechanical model of the lumbar torso was used in this effort. The model supersedes earlier efforts by others in that it contains recent data on the torso muscle centroids and lines of action (McGill et al., 1993; Moga et al., 1993), includes compressive preload effects on the reactive spinal bending moments (Adams et al., 1991), and accounts for the effects of pelvic rotation on torso muscle line of action and length estimations. It also integrates the optimization method of Bean et al. (1988) to estimate the tensions of the torso musculature based on the load moment at the L4/L5 vertebral interspace. The muscle parameters estimated using the biomechanical model were used in conjunction with myoelectric recordings to estimate the length-tension and velocity-force parameters of the torso extensors.

A widely used function (Eq. 1) was used to relate the EMG data with muscle force. The function maps the myoelectric to tension relationship using both length-tension and velocity of shortening factors in a product relationship (Abbot and Wilkie (1953); Bahler et al. (1967); Olney and Winter (1985)).

$$F_i = \mathrm{Gain}_i * \mathrm{EMG}_i * f(\mathrm{l}) * f(\mathrm{v}) + e_i \quad (1)$$

where F_i is the tension produced by muscle_i, EMG_i is the normalized integrated EMG for muscle_i, Gain_i is the muscle_i EMG to tension scaling, $f(\mathrm{l})$ is the muscle length compensation function, $f(\mathrm{v})$ is the muscle velocity compensation function, and e_i are the error residuals.

A technique was developed to estimate the length-tension and velocity-force functions using EMG data from the erector muscles captured under various controlled motion and tension conditions.

Motions at three velocities (10, 20 and 30 deg/sec) were recorded with emphasis on three "zones" of torso flexion (0, 22.5 and 45 degrees). Myoelectric data collected were normalized to each of these ranges. All of the data collected throughout the range of a trial were used to develop the muscle EMG-tension relationships. During constant speed trials, the velocity modulator function $f(\mathrm{v})$ could be assumed constant allowing simplification of the model (Eq. 2):

$$f(\mathrm{l}) = F_i / (\mathrm{Gain}_i * \mathrm{EMG}_i) \quad (2)$$

Once an estimate of the length modulating function was obtained, the velocity tension function could be investigated using trials within a specific zone (i.e. 22.5 degrees of torso flexion) at the slow, medium and fast trial rates. Thus, having an estimate of $f(\mathrm{l})$ allows:

$$f(\mathrm{v}) = F_i / (\mathrm{Gain}_i * \mathrm{EMG}_i * f(\mathrm{l})) \quad (3)$$

Experimental Methods

Five male subjects participated in the study (mean (SD) age: 29.6 (4.9) years; mean height: 179.6 (8.3) cm; mean weight: 73.7 (8.4) kg). Subjects stood on a force platform with their ankles, knees and pelvis attached via straps to a rigid fixture bolted into the top of the platform (Figure 1). The subjects performed sagittal plane lifts using weights of 153 and 247N while their postures were dynamically measured using their postures were dynamically measured using a 3-D sonic coordinate measuring system (Raschke, 1994). Myoelectric signals were recorded from the main erectors (erector spinae and latissimus dorsi), external oblique and rectus abdominus (right side).

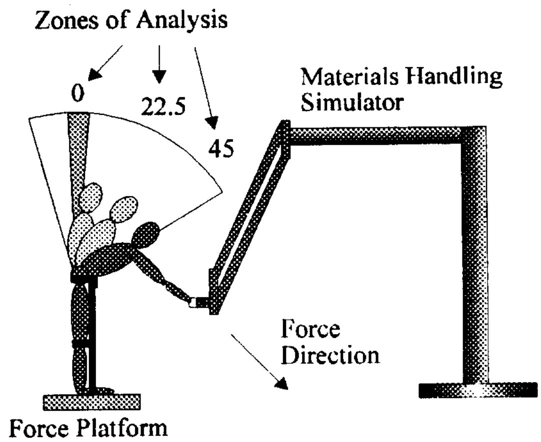

Figure 1. Three zones of torso flexion used to find the length-tension and force-velocity relations.

Procedure

The subjects were asked to perform sagittal plane lifts beginning in a stooped posture and ending in a neutral upright posture.

The normalized force data (Eq. 2) from the 22.5 degree zone were plotted with respect to the muscle length to reveal the length-tension relationships. The optimization method incorporated in the biomechanical model was used to estimate the muscle forces for this procedure. Least-squares analysis was used to find the coefficients of the linear model for each subject and to test the goodness of fit. These coefficients were compared against those found using the data from the 0 and 45 degree zones for validation. The error residuals were inspected to validate the use of a linear model. Once the length modulators were determined for each subject and muscle, the velocity modulator was estimated using the slow, medium and fast movement data from the 22.5 degree flexion zone.

Results

The least squares analysis and inspection of the error residuals supported the use of linear models to describe the erector spinae and latissimus length-tension relationships for all subjects (Figure 2; ES average $R^2 = 0.47$, std. err. = 0.15; LD average $R^2 = 0.33$, std err. = 0.21N). Antagonistic muscle activity was minimal and exhibited no trend.

For the velocity data, the exponential model of the form predicted by the Hill (1938) equations was found to be only a very modest improvement over a linear model, explaining maximally an additional two percent of the observed myoelectric variation for these data (Figure 3).

Analysis of the length-tension and velocity-force modulator effects on the myoelectric to muscle tension relationship revéaled a substantial improvement when the length modulator was included ($R^2 = 0.35$ vs. $R^2 = 0.66$) and a lesser, but significant improvement with the additional inclusion of the velocity modulator ($R^2 = 0.66$ vs. $R^2 = 0.71$). The standard error of the regression fit was reduced by 50% (218N vs. 110N) through the inclusion of both modulators.

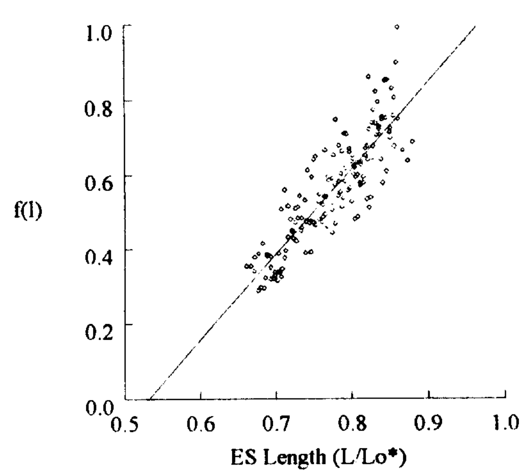

Figure 2. Data for a representative subject showing the length-tension relationship [f(l)]. The data indicates a increasing tension capability as the muscle length

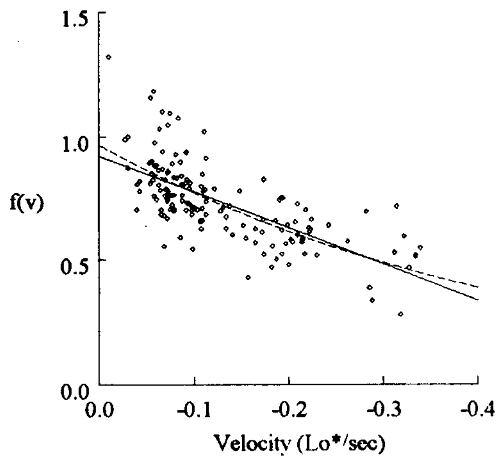

Figure 3. The length and tension normalized EMG plotted against the erector spinae contraction velocity, revealing the velocity modulator [*f*(v)] for a representative subject. Both linear (solid line: r^2 = 0.48) and exponential (dashed line: r^2 = 0.50) model fits are shown.

Discussion

The data supported the hypothesis that the length-tension relationships of the extensor muscles follow the linearly increasing strength capability trend toward full flexion. For the velocity-force relations, an exponential decrease in tension capability with increased contraction velocity was observed, supporting the *status quo* hypothesis.

Characteristics of muscles exhibiting linear length-tension relations may present a mechanism for low back pain. These muscles, such as the tibialis anterior muscle of the cat, have been shown to produce up to 25% of their maximum active force when stretched to the point of aponeurosis tissue failure (Gareis et al., 1992). Assuming that the erector spinae has similar active force characteristics to the tibialis anterior, individuals in extreme flexion with their erectors fully stretched may have sufficient active tension capability in these muscles to cause injury to the muscle aponeurosis, leading to back pain.

The approach used in the study to obtain the length-tension and velocity-force relations seems to be applicable to muscles that cannot be evaluated using invasive methods, and resulted in fairly robust relationships. The results are expected to enhance the prediction accuracy of dynamic muscle tensions in EMG-assisted biomechanical models, and aid in the interpretation of myoelectric values in the validation of predictive models. Both of these modeling approaches are currently being used in the investigation of low back pain.

Acknowledgments

This study was performed with the financial assistance of Coca Cola, USA and Ford Motor Company.

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