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# Modification of an EMG-assisted biomechanical model for pushing and pulling

E.W. Theado, G.G. Knapik, W.S. Marras*

*The Biodynamics Laboratory, The Ohio State University, 210 Baker Systems, 1971 Neil Avenue, Columbus, OH 43210, USA*

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## Abstract

Pushing and pulling tasks using carts and material handling devices have become more prevalent in occupational environments in an attempt to reduce the musculoskeletal risks associated with lifting. However, little change in low back disorder rates have been noted as tasks change from lifting to pushing and pulling indicating that we do not understand the mechanics of pushing and pulling well. Biomechanical assessments of pushing and pulling tasks using person-specific biologically assisted models offer a means to help understand how the spine is loaded under pushing and pulling conditions. However, critical components of these models must be adjusted so that they are sensitive to the different physiologic responses in the torso muscles expected during pushing and pulling compared to lifting tasks.

The objective of this study was to modify an electromyography (EMG)-assisted biomechanical model designed to evaluate lifting tasks so that it can better represent the biomechanical forces expected during pushing and pulling tasks. Several key modifications were made. Based upon a literature review, changes in muscle cross-sectional area and muscle origins and insertions were made to better represent the geometry of the torso muscles. It was also necessary to adjust the length–force and velocity–force muscle relationships. Empirically derived length–force and velocity–force relationships were developed to independently represent the flexor and extensor musculature. These modifications were then systematically incorporated into the model.

The model was exercised over several pushing and pulling conditions to assess the effect of these modifications on its ability to predict externally measured spinal moments. Results indicated that the alterations made to the preexisting EMG-assisted model resulted in acceptable model performance for pushing, pulling, and lifting activities.

## Relevance to industry

The use of carts and material handling devices has become increasingly prevalent in industry, though little research has been done to examine the body's response. The modifications made to the biomechanical model would enable its use in the evaluation and design of material handling devices and pushing and pulling tasks.



*Keywords:* Spinal loading; Electromyography; Push and pull; Biomechanical modeling

## 1. Introduction

Pushing and pulling are among the most common methods of manual material handling (MMH) seen in industry. They typically consist of tasks using carts (de Looze et al., 2000; Schibye et al., 2001) and material handling devices (MHDs) such as hoists or lifts (Nussbaum and Chaffin, 1999; Resnick and Chaffin, 1997). While these devices are employed to eliminate lifting and lessen the load on workers, studies have reported that 9–18% of lower back injuries are associated with pushing and pulling (Hoozemans et al., 1998; Shoaf et al., 1997). MMH tasks that involve pushing and pulling have been shown to invoke a significantly different physiologic response when compared to lifting tasks. First, during pushing and pulling, flexion and extension moments are applied to the spine, whereas lifting exertions expose the spine exclusively to extension moments. Second, there are smaller torso angles observed in pushing and pulling relative to lifting.

*Corresponding author. Tel.: +1 614 292 6670; fax: +1 614 292 7852.
*E-mail address:* marras.1@osu.edu (W.S. Marras).

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As a result of theses differences, pushing and pulling tasks need to be investigated to determine the physiological load response these tasks have on the industrial working population.

The goal of this study was to modify the lifting model developed by Marras and colleagues at The Ohio State University (Granata and Marras, 1993, 1995; Marras et al., 1999, 2001a; Marras and Granata, 1995, 1997a, b; Marras and Sommerich, 1991a, b; Reilly and Marras, 1989) to assess the loads that are imposed upon the lumbar spine during flexion and extension moment exertions characteristic of pushing and pulling.

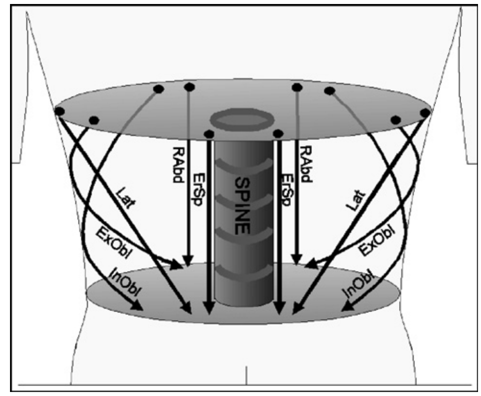


Fig. 1. Physical representation of model logic.

## 2. Basic model structure

The key to understanding how the lumbar spine is loaded during different activities is to understand how the internal forces (primarily muscles) within the trunk react to these different tasks. Many previous models attempted to predict how muscles respond to a task; however, these models must adopt simplifying assumptions, or optimization algorithms in order to estimate muscle activity levels. None of these approaches have been able to predict muscle activities under realistic, dynamic conditions. Instead of predicting how muscles would respond biologically assisted models directly monitor the trunk muscle responses to physical loading conditions and use this information as input to a biomechanical model so that the effects of realistic muscle coactivation can be considered in defining tissue loading. It is common for biologically assisted models to employ electromyography (EMG) as a means to monitor muscle activity.

Our EMG-assisted dynamic model is unique in that it is *person specific* in terms of: (1) anthropometry (muscle location and size), (2) body mass characteristics, (3) subject motion (inputs trunk as well as limb motion), and (4) muscle activities. The model structure is multi-dimensional and is capable of considering the dynamic response of the individual. Trunk moments and tissue loads are derived from dynamic muscle force vectors and internal trunk muscle moment arms. Our model currently considers the 10 power-producing muscle equivalent vectors to and their relationship to represent trunk anatomy and mechanics (Dumas et al., 1991; McGill et al., 1988; Schultz and Andersson, 1981; Yoo et al., 1979) (see Fig. 1). Trunk muscle vector directions have been derived via MRI studies (Jorgensen et al., 2001). Originally, muscle fibers sampled by EMG were monitored via intramuscular electrodes (Marras et al., 1984); however, years of testing in our laboratory have permitted us to obtain equivalent sensitivity using surface electrodes.

The EMG derived from the 10 power-producing muscles is assumed to be representative of, and linearly related to the net muscle force. Some authors (Yoo et al., 1979; Lippold, 1952, 1970; Moritani and deVries, 1978; Moritani et al., 2005) believe surface EMG is linearly related to voluntary isometric joint torque, whereas others (Komi and Vitasalo, 1976; Vredenbregt and Rau, 1973; Zuniga and Simons, 1969) believe the relation is strictly non-linear. Hof and van den Berg (1977) explained this paradox by demonstrating that EMG is linearly related to muscle force, but is non-linearly related to joint torque due to coactivity. Since EMG-assisted models account for muscle coactivity, a linear assumption is reasonable.

The model employs kinematic information (measured from the specific subject) and anthropometrically scaled musculoskeletal anatomy to determine the muscle vector directions. Processed (rectified and averaged) EMG data and calibrated muscle stress values (gains) are combined with muscle size, length, and contraction velocity to determine the force magnitudes. Multi-dimensional, dynamic, trunk moments, and spinal loads are computed from muscle forces and moment arm vectors.

The mechanics of this model can be visualized as two "plates" (one located at the sacrum and one at T12) that move relative to one another (Fig. 1). The model is free dynamic in that it permits each muscle's orientation, length, and velocity to vary individually throughout the exertion. This makes it possible to constantly update muscle position, length, and velocity. Muscle origins are assigned a three-dimensional orientation location relative to the spinal axis, coplanar with the iliac crest. Muscle insertions are located coplanar with the 12th rib. Muscle forces are represented as vector quantities between their two endpoints.

Dynamic motions of the trunk are recorded from the subject using a lumbar motion monitor (LMM) placed over the back and appropriately modeled via the relative dynamic orientations of the two mathematical planes or "plates." Therefore, as the subject moves, the mechanical advantage of each muscle vector is adjusted so that their contribution to tissue loading is constantly updated. Specific vertebral body orientations are also instantaneously defined via the LMM position (Splittstoesser, 2001). Since the origins and insertions of the 10 power-producing trunk muscles are fixed on the surface of the thoracic and illiac "plates," movement of these plates relative to one another are representative of dynamic motions of the trunk (measured by the LMM) and capable of generating three-dimensional motion in the muscle. Using this approach, muscle orientations and relative muscle lengths are permitted to change throughout a movement, realistically representing each muscle's changing mechanical advantage during a task. Muscle vector directions, lengths, and velocities are continuously determined from the instantaneous positions and motions of the muscle endpoints.

Spinal loads are calculated from the vector sum of the 10 muscle forces (Eq. (1)). Muscle-generated moments ($M$) about the spinal axis are predicted from the sum of vector products combining dynamic tensile forces ($F$), and moment arms ($r$) of each muscle, ($j$).

$$\vec{M} = \sum_{j} \vec{r}_j \times \vec{F}_j \tag{1}$$

The tensile force generated by each muscle, $j$, is described (Eq. (2)) by the product of normalized EMG, muscle cross-sectional area, a gain (maximum muscle stress) factor representing muscle force per unit area, and modulation factors describing EMG and force behavior as a function of the length [$f(\text{Length}_j)$], and velocity [$f(\text{Vel}_j)$] of the muscle:

$$\text{Force}_j = \text{Gain}\,\frac{\text{EMG}_j(t)}{\text{EMG}_{\text{Max}\,j}}\,\text{Area}_j\, f(\text{Vel}_j)\, f(\text{Length}_j) \tag{2}$$

EMG data are normalized relative to myoelectric maxima from each muscle. This is necessary to remove possible analytical errors related to electrode placement, skin abrasion, flesh resistance, muscle fiber density, and electronic channel differences. Normalized EMG data are multiplied by their respective muscle cross-sectional areas to account for the relative force-generating capacity of each muscle since maximum muscle force is directly related to cross-sectional area. Our previous investigations have determined how the muscle cross-sectional area can be reasonably derived as a function of anthropometric measures (Marras et al., 2001b). The force–length relationship and force–velocity relationship shown in Eq. (2) have been derived experimentally. Gain is calibrated for an individual subject and must fall within the physiologic range.

## 3. Model modifications

Model modifications consisted of fine-tuning trunk muscle representations and functions to better represent their involvement in complex exertions such as pushing and pulling activities. Application of force in the horizontal direction, as is the case when pushing, requires a flexion moment about the spine. Therefore, when predicting flexion in addition to extension trunk moments, it is critical that the flexor muscles are represented correctly. Under simple lifting conditions, flexion moments are less critical to model performance since the exertion is dominated by extension moments.

Several modifications to the basic model were required in order to make the model sensitive to pushing and pulling activities. First, it is expected that the orientation and mechanical advantage of the flexor muscles relative to the spine would change within the trunk during flexion since the lumbar curve changes during flexion activities. Thus, an adjustment was needed to the flexor muscle vector orientations described in Fig. 1. The lifting model uses data from a supine MRI study to assign the muscle origins and insertions (Jorgensen et al., 2001). These origins and insertions define the moment arms of the 10 trunk muscles, which in turn have a profound influence on defining spine tissue loads. McGill et al. (1996) used ultrasound measurements to determine that the rectus abdominis moment arms in the supine position are different than those in the prone position. This investigation found that the moment arms increased by 30% when activated in the prone position compared to the relaxed supine position. These findings were recently confirmed using upright MRI imaging (Jorgensen et al., 2005). In addition, a dissection study (McGill, 1996) revealed that the majority of the external oblique muscle connects to the linea semilunaris, which is the sheathing of the rectus abdominis. These results suggest that the external oblique muscle should have the same moment arms as the abdominals. As a result, the moment arm location of the rectus abdominus in our model was increased by 30% in the model and the external oblique moment arm was equated with the rectus abdominus.

Second, it is also imperative to represent muscle area correctly so that one can determine the amount of muscle area that is contributing to the moment imposed about a particular joint. The latissimus dorsi muscle can adduct and extend the shoulder and also acts in extension on the lumbar spine. Bogduk et al. (1998) performed a gross dissection on human cadavers to determine how much of the cross-sectional area of the latissimus dorsi contributes to the extension of the lumbar spine. They concluded that 43% of the muscle fibers could produce an extension moment about the L5/S1 intervertebral disk. In order to account for this adjustment, the model was changed to account for the smaller latissiumus dorsi cross-sectional area.

Third, the flexor muscle's length–force relationship (used in Eq. (2)) must be adjusted for flexion. The original model assumed a similar length–force relationship for all muscles. However, it is possible and likely that flexion trunk muscles would have a different length–force relationship compared to extension muscles. The optimal length of the muscle is the length at which it is able to produce the maximum amount of active force and is the anchor point for the length–force relationship. This becomes an extremely important issue with the current model of the erector spinae because the optimal length occurs when flexed forward 20°. In pushing and pulling there is relatively little back angle when compared to lifting tasks, which puts the erectors at a mechanical disadvantage. A review of literature revealed that when modeling the trunk muscles, the anatomical upright position is a valid assumption of the optimal muscle length. McGill (1991) examined kinetic parameters in the trunk muscles and cited the resting length of all the trunk muscles in the upright position. Biomechanical models of the lumbar spine that use a stability approach (Granata and Wilson, 2001) and an optimization approach (Daggfeldt and Thorstensson, 2003; Nussbaum and Chaffin, 1996) both cite the resting length as being in the upright anatomical position, so this change was also incorporated into the model.

A muscle's length and contraction velocity have a profound effect on the ability to produce force and are referred to as force–length and force–velocity modulation factors. In the original model, all 10 muscles are modeled using modulation curves from the most recent EMG-assisted biomechanical model. A muscle's force–length and force–velocity relationships have been shown to be highly dependent on the architecture of that particular muscle (Narici, 1999; Lieber and Friden, 2000). It was reported that the force–length curve of a pennate muscle (erector spinae) would have a sharper peak but a smaller active muscle range than a parallel fibered muscle (rectus abdominis).

### 3.1. Empirical determination of force–length and force–velocity relationships

Two exertions were designed to isolate the flexors and extensors torso muscles in order to empirically derive two separate force–length and force–velocity modulation factors based on external moments and muscle kinematic values. Twelve subjects (six males and six females) were tested to derive the force–length, and force–velocity curves comparing flexion and extension exertions. All subjects were healthy University aged subjects without a history of low back pain.

The extension exertions consisted of a typical free dynamic sagittal lifts of approximately 6.8 kg with a lift origin of waist height and a moment arm of one-half meter from spine.

The flexion exertions required the subjects to support their legs and buttocks in a supine posture while freely suspending their torso parallel to the floor (Fig. 2). The exertion began with $0^{\circ}$ of trunk angle followed by approximately $10^{\circ}$ of hyperextension. Once in these starting postures, the subjects flexed their torso's until they reached an upright seating posture.

The extension exertions examined the dynamic ratio of measured moment to predicted moment (using the EMG-assisted model) in order to derive the force–length relationship. This ratio was plotted as a function of the averaged normalized muscle length of the right and left erector spinae muscles. The ratio represents the amount of actual extension moment that is inappropriately estimated by the predicted extension moment based upon the muscle activity. The offset is attributed to the length–strength relationship in the processed EMG. Each subject was analyzed using this method to produce individual length–strength relationships for extension and plotted on one figure. A third-order polynomial trend line was used to represent a general length–strength relationship for all subjects.

Similarly, for the flexion exertions the EMG-assisted model was used to output trunk kinematics and were used to obtain a simulated external moment and derive the flexor muscles length–strength relationship. The external moment was simulated instead of measured because the flexion task needed to limit the amount of activity of the extensor muscles during the exertion.

The extensor and flexor force–length relationship curves are shown in Figs. 2–4, respectively.

The newly derived force–length modulation factor was then applied to the model and the data were reanalyzed. The dynamic ratio of the measured or simulated moment to the predicted moment was plotted against the normalized contraction velocity of the erector spinae for extension and the rectus abdominus for flexion. The ratio represents

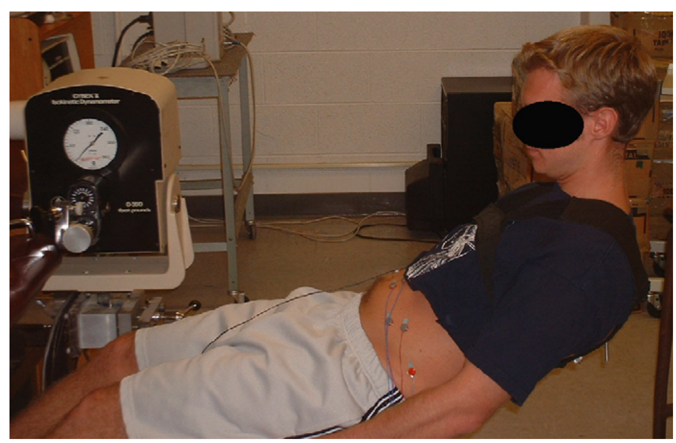

Fig. 2. Sit-up exertion used to derive length–strength and force–velocity for flexor muscles.

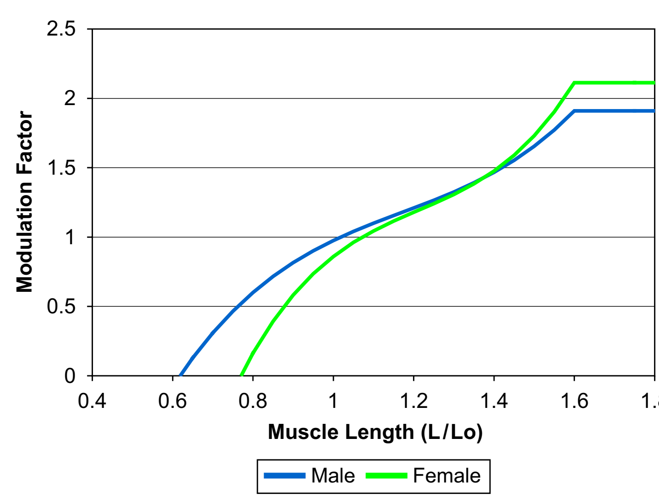


Fig. 3. Extensor muscle group force–length relationship.

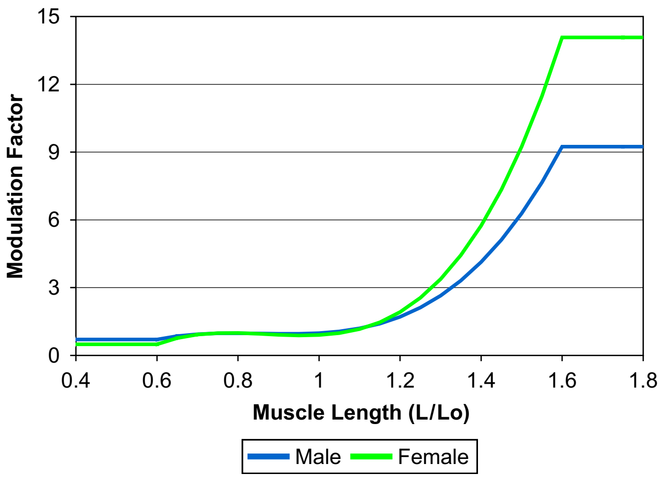


Fig. 4. Flexor muscle group force–length relationship.

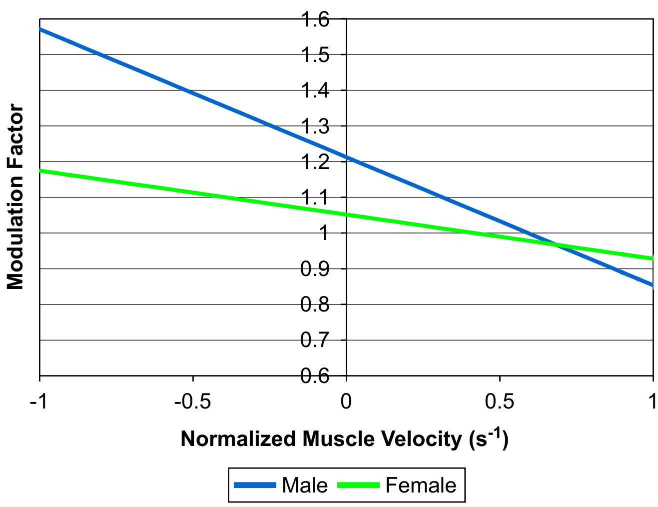


Fig. 5. Extensor muscle group force–velocity relationship.

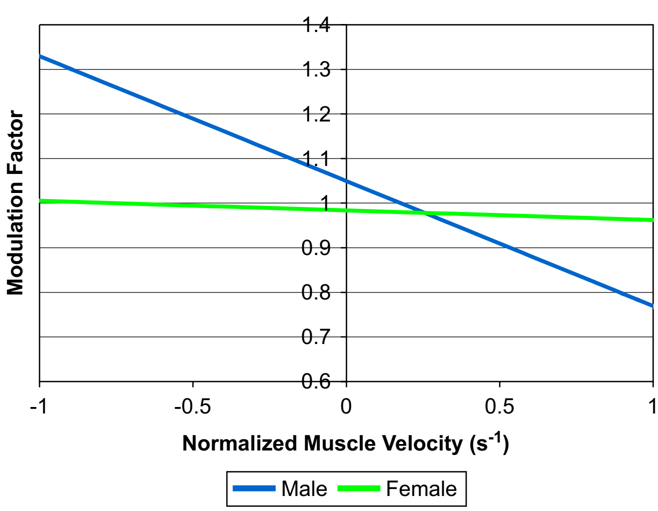


Fig. 6. Flexor muscle group force–velocity relationship.

the amount of measured moment that is inappropriately estimated by the predicted moment from the muscle activity. The relative difference in moments was attributed to the force–velocity effect in measured EMG. Each subject was analyzed using this method to produce individual force–velocity relationships for flexion and extension. These individual force–velocity relationships were then plotted on one graph for both flexion and extension. A first order polynomial trend line, the same order as the current model, was used to represent a general force–velocity relationship for flexion and extension. These relationships are shown in Figs. 5 and 6 for the extensor and flexor muscles, respectively.

## 4. Model performance

A study was performed to assess the effects of the model modifications upon model performance. Ten males (mean age 24.5, SD = 3.24) and 10 females (mean age 22.7, SD = 2.36) were recruited to test the model performance in sagittal plane lifting compared to pushing and pulling. Subject stature (male average = 180.5 cm, SD = 7.32; female average = 165.4 cm, SD = 4.2) and weight (male average = 72.7 kg, SD = 10.2; female average = 57.1, SD = 8.2) were typical of a University population of subjects.

The experimental tasks consisted pushing or pulling upon handles supported by a rigid structure mounted on an

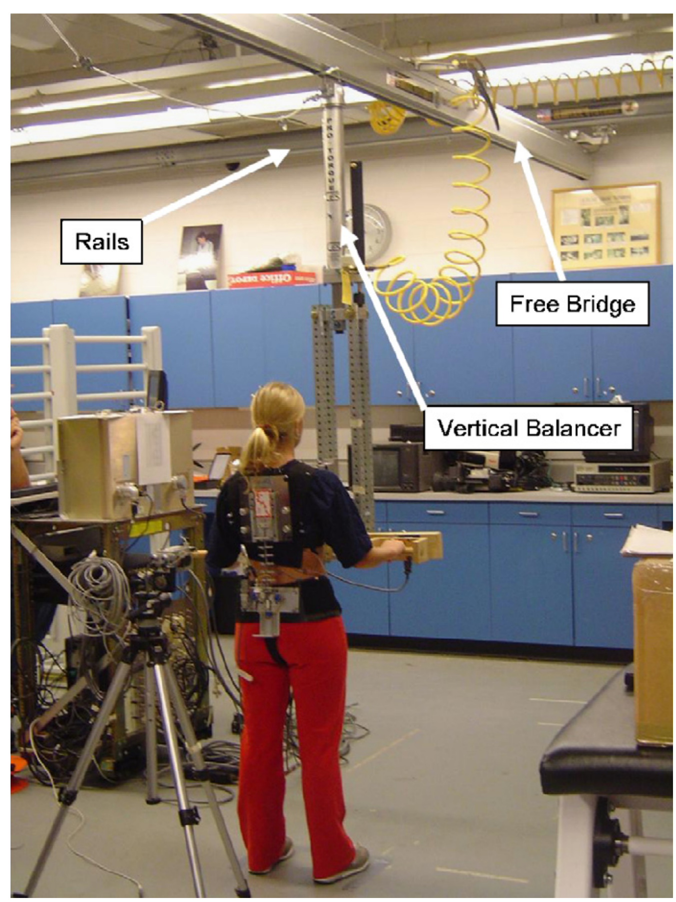


Fig. 7. Experimental apparatus used to evaluate pushing and pulling model performance.

overhead rail (Fig. 7). The handles were instrumented with force transducers that were able to monitor the force applied in three dimensions as well as the torque produced around each handle in all three directions. The handles were adjusted between 50% and 80% of subject stature and subjects produced handle forces equivalent to 20%, 30%, and 40% of subject body weight. In addition, subjects were asked to lift a 11.4 kg box from knee to waist height (within the sagittal plane) while standing on a force plate. This condition represented a "sagittal lift" and was used to compare the original lifting model performance (Marras and Granata, 1997b) to pushing and pulling performance.

Subjects began each exertion with the feet together standing erect with their hands at their sides in order to take a neutral reference value and to zero the hand transducers. Subjects were then instructed to place their hands around the handles without touching them until instructed to begin the task. The subjects were instructed to start with a specific initial arm posture at the beginning of each trial in order to remain consistent with the initial conditions in the push–pull model. During the experiment, subjects were instructed to push and pull at a speed that was fast yet comfortable. No restrictions were placed on how to step or stride length. Subjects were provided rest periods between each trial.

Instrumentation has been described in depth in previous studies (Granata and Marras, 1995; Marras et al., 2001a; Marras and Granata, 1995, 1997a). The electromyographic activity of the 10 trunk muscles used in the model were monitored as was the kinematic activity of the trunk and the kinetic forces generated at the hands (for pushing and pulling) or at the feet (for sagittal lifting conditions).

Model performance was evaluated by comparing the measured torque about the spine via the hand transducers or force plate to the toque predicted via the EMG-assisted biomechanical model. The measure of performance selected for comparison consisted of the average absolute error (AAE) between the measured toque and the predicted torque during the exertion.

The AAE values were not significantly different between pushing and pulling (12.71 and 12.20 N m, respectively). For comparison, the AAE resulting from the sagittal lifting conditions was 7.45 N m. Hence, while the model performed extremely well for lifting activities, it also performed acceptably well for pushing and pulling activities.

## 5. Discussion

This effort has explained how biologically or EMG-assisted models use information about the biomechanical functioning of the subject to estimate loads upon the lumbar spine's tissues. Previous EMG-assisted models have been concerned primarily with lifting activities. It has not been uncommon to observe these models being applied to more complex exertions. However, as these models are employed to assess more complex exertions, it is important that the model parameters be adjusted to represent the muscle architecture, force–length, and force–velocity relationships inherent in the muscles that are actively involved in the exertions of interest. This effort has described how these measures were adjusted for pushing and pulling exertions.

Several important modifications were necessary to describe pushing and pulling exertions. These modifications included modification to the origin and insertions of the trunk's flexor muscles, as well as the derivation of independent flexor and extensor force–length and force–velocity relationships. These differences are shown in Figs. 3–6 indicating that previous assumptions that all muscles had similar muscle relationships were inappropriate assumptions. This effort has shown how differences in these parameters for flexion and extension of the torso are necessary for proper model functioning. This work indicates that both the force–length and force–velocity relationships must be adjusted for flexion and extension in order to maintain model integrity.

The model performance study has indicated that the changes described resulted in a very accurate model for pushing and pulling activities as well as lifting. Thus, these model modifications provide a significant advancement in model development and will help us more accurately assess the effects of physical activities upon spine tissue loading.

## 6. Conclusions

This research was performed to investigate the impact of changes in muscle geometry and the length–strength and force–velocity relationships on modeling accuracy and spinal loads as it pertains to pushing and pulling tasks. The muscle geometry was altered by incorporating the following changes that were found in the literature: increasing the rectus abdominis moment arm by 30%, equating the external oblique origin to the rectus abdominis, reducing the area contribution of the latissimus dorsi to include only 43% of the total muscle area, and setting all the trunk muscle's optimal lengths to occur in the anatomical upright position. With the updated muscle geometry, the length–strength and force–velocity relationship for the extensors (erector spinae, latissimus dorsi, and internal oblique) and the flexors (rectus abdominis and external oblique) were empirically derived.

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