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A biologically-assisted curved muscle model of the lumbar spine: Model structure

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ABSTRACT

Background: Biomechanical models have been developed to assess the spine tissue loads of individuals. However, most models have assumed trunk muscle lines of action as straight-lines, which might be less reliable during occupational tasks that require complex lumbar motions. The objective of this study was to describe the model structure and underlying logic of a biologically-assisted curved muscle model of the lumbar spine.
Methods: The developed model structure including curved muscle geometry, separation of active and passive muscle forces, and personalization of muscle properties was described. An example of the model procedure including data collection, personalization, and data evaluation was also illustrated.
Findings: Three-dimensional curved muscle geometry was developed based on a predictive model using magnetic resonance imaging and anthropometric measures to personalize the model for each individual. Calibration algorithms were able to reverse-engineer personalized muscle properties to calculate active and passive muscle forces of each individual.
Interpretation: This biologically-assisted curved muscle model will significantly increase the accuracy of spinal tissue load predictions for the entire lumbar spine during complex dynamic occupational tasks. Personalized active and passive muscle force algorithms will help to more robustly investigate person-specific muscle forces and spinal tissue loads.

1. Introduction

Since the 1970s, biomechanical models have been developed to estimate spine tissue loads to quantify the risk of spine disorders for workers in occupational environments. These spine tissue load estimations have usually been compared with the tolerance limit of intervertebral disk endplates, and this relationship has been used to indicate the portion of the population exposed to risk of spine tissue damage in specific work places.

Early on, a static single-equivalent-muscle model was developed to evaluate simple lifting tasks (Chaffin, 1969). However, this model assumed that static postures were representative of lifting movements, focused on trunk extensor muscles exclusively, and assumed those muscles could be represented as a single equivalent trunk muscle group. Only a single component of spine tissue load (compression) was originally calculated, and these models typically assumed that no muscle co-activation occurred during lifting tasks.

Static multiple-muscle models were later developed to account for the various contributions of multiple muscles surrounding the spine during lifting (Schultz and Andersson, 1981). Ten trunk muscles and intra-abdominal pressure were included. This model also considered shear forces as well as compression loads imposed on the spine. Although this approach investigated the effect of multiple muscles on spinal loads, it assumed no antagonistic muscle activity existed during lifting. This assumption was assumed to be appropriate for static lifting exertions but not for dynamic lifting exertions that commonly required greater co-contractions of trunk muscles (Marras et al., 1984). In addition, studies found that ignoring co-contraction of muscles could underestimate spinal loads by up to 70% (Granata and Marras, 1995a,b).

Biologically-assisted models were developed to account for the co-activation of multiple muscles (McGill and Norman, 1986). They

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measured electromyography (EMG) data as an input to assess muscle coactivity, but were limited to validation during static exertions only. Static stability models were also introduced (Cholewicki and McGill, 1996; Granata and England, 2006), but they were not applicable to dynamic stability behavior.

Dynamic biologically-assisted models were also developed in an effort to account for muscle co-activation during dynamic exertions (Arjmand et al., 2010; Gagnon et al., 2011; Granata and Marras, 1993; Marras and Sommerich, 1991a,b; McGill, 2004; Van Dieen and Kingma, 2005). Some of these advanced models attempted to assess the person-specific spine tissue loads of individual workers during a wide range of dynamic occupational tasks, such as lifting (Arjmand et al., 2011; Marras et al., 2004), pushing/pulling (Knapik and Marras, 2009; Marras et al., 2009), and carrying (Rose et al., 2013; Schibye et al., 2001).

Although advanced models were able to assess biological responses of multiple trunk muscles during dynamic occupational tasks, they assumed the muscle lines of action could be represented as straight-line vectors (Granata and Marras, 1993; Marras and Sommerich, 1991a,b; McGill and Norman, 1986). "Straight-line" muscle models have worked reasonably well in relatively simple tasks that require only a small range of lumbar motions, but this assumption could be less reliable when applied to complex asymmetric lumbar motions that commonly occur in the workplace (Marras et al., 1993). These improper muscle force lines of action assumptions could affect the model's accuracy of predicting three-dimensional spine tissue loads.

Another important limitation of most models is that they have rarely considered the individual variability of muscle properties, such as muscle force-length and force-velocity relationships. Most models assume the same relationship of muscle force-length and force-velocity for all subjects (Christophy et al., 2012; Ghezelbash et al., 2015; Van Dieen and Kingma, 2005), or do not account for these relationships at all (Daggfeldt and Thorstensson, 2003; de Zee et al., 2007). However, previous literatures reported age-related changes of muscle force-length and force-velocity relationships (Thelen, 2003), and variations of optimal sarcomere lengths in humans (Lieber et al., 1994; Walker and Schrodt, 1974). Various slopes and shifts of the muscle force-length and force-velocity relationships directly affect the magnitude and temporal variability of the muscle forces as a function of the changes of muscle length and muscle velocity of individuals. Considering this physiological variability of muscle properties would help to estimate more person-specific muscle forces.

In addition, only a few spine models clearly have distinguished the role of active and passive muscle force in force calculation algorithms (Ghezelbash et al., 2015; Van Dieen and Kingma, 2005). If models only rely on active muscle force components during muscle activities, it is difficult to precisely assess the spinal loads during lumbar flexion relaxation which involves minimal muscle activity but significant passive force (Adams and Hutton, 1986; Ghezelbash et al., 2015; Hajihosseinali et al., 2014).

In order to overcome these issues, a personalized biologically-assisted curved muscle model was developed in the current study. Several curved muscle models have previously been developed for both the cervical spine (Kruidhof and Pandy, 2006; Suderman and Vasavada, 2012; Vasavada et al., 2008) and the thoracic/lumbar spine (Arjmand et al., 2006; Gatton et al., 2001; Stokes et al., 2010; Van Dieen and Kingma, 2005). However, most of these models have not been validated during dynamic exertions and personalization of muscle parameters was seldom considered. In addition, only extensor or only oblique abdominal muscles were generally treated as curved muscles (Arjmand et al., 2006; Stokes et al., 2010). Herein, we developed an approach that overcomes many of the previously discussed limitations in order to better predict spinal tissue loading during complex dynamic occupational tasks that is personalized to each individual. A specific example illustrating the model's fidelity is presented.

2. Model development

2.1. Goals

The goals of this effort were two-fold:

1) Develop accurate three-dimensional curved muscle geometry in the model based on Magnetic Resonance Imaging (MRI)-derived muscle moment-arms and physiological cross-sectional areas (PCSAs) as a function of anthropometric measures.
2) Develop a personalized biologically-assisted muscle force algorithm to account for individual variation of trunk muscle properties including both active and passive muscle gains and muscle force-length and force-velocity relationships.

2.2. Benchmark model structure

The benchmark model structure is consisted of a biologically-assisted straight-line muscle model of the lumbar spine. This model utilized individual anthropometry, kinematics, kinetics, and biological muscle activities to predict the three-dimensional spinal loads during dynamic exertions and has been well validated across different types of occupational tasks (Dufour et al., 2013; Granata and Marras, 1995b, 1993; Knapik and Marras, 2009; Marras and Granata, 1995, 1997a, 1997b; Marras and Sommerich, 1991a, 1991b; Marras et al., 2009, 2004, 2001a; Rose et al., 2013; Theado et al., 2007). The straight-line muscle model represented the trunk muscles as ten force vectors attached between the upper and lower torso. It calculated the spinal loads at multiple lumbar disk levels based on the summation of these multiple muscle force vectors and the inertial contributions of different body segments.

2.3. Curved muscle model structure

Fig. 1 shows the conceptual flow diagram of the biologically-assisted curved muscle model. It illustrates the internal components of the models from model inputs to model outputs, and their interactions. In particular, elements highlighted with a dotted line represent how the curved muscle representation systematically affects multiple components in the model.

With regards to the model inputs, positional data from an optical motion capture system were used to determine the location of the spine and other body segments relative to a force plate. Kinematic data including angle, angular velocity and angular acceleration of the trunk and whole body were used to calculate the trunk muscle length, muscle velocity, and gravitational moments of each body segment. For kinetic data, measured three-dimensional external force and torque from force transducers was used to calculate spinal moments at multiple lumbar levels. The model also utilized EMG data from five pairs (10 muscles) of major power-producing trunk muscles including the latissimus dorsi, erector spinae, rectus abdominis, external oblique, and internal oblique to calculate the biologically-assisted muscle force. Muscle forces were also modulated based on person-specific parameters describing the relative active and passive strength of the muscle, as well as relationships between force and the instantaneous length and velocity of each muscle. Subject anthropometric data including height, body mass, trunk width, and trunk depth measures, and demographic information such as age and gender were required to predict the personalized curved muscle geometry for each individual. An MRI database of muscle moment-arms and PCSAs of the ten trunk muscles at multiple thoracic/lumbar levels was used to develop the precise anatomical curved muscle geometry (Jorgensen et al., 2001; Marras et al., 2001b).