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Short communication

A model-based approach for estimation of changes in lumbar segmental kinematics associated with alterations in trunk muscle forces

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ABSTRACT

The kinematics information from imaging, if combined with optimization-based biomechanical models, may provide a unique platform for personalized assessment of trunk muscle forces (TMFs). Such a method, however, is feasible only if differences in lumbar spine kinematics due to differences in TMFs can be captured by the current imaging techniques. A finite element model of the spine within an optimization procedure was used to estimate segmental kinematics of lumbar spine associated with five different sets of TMFs. Each set of TMFs was associated with a hypothetical trunk neuromuscular strategy that optimized one aspect of lower back biomechanics. For each set of TMFs, the segmental kinematics of lumbar spine was estimated for a single static trunk flexed posture involving, respectively, 40° and 10° of thoracic and pelvic rotations. Minimum changes in the angular and translational deformations of a motion segment with alterations in TMFs ranged from 0° to 0.7° and 0 mm to 0.04 mm, respectively. Maximum changes in the angular and translational deformations of a motion segment with alterations in TMFs ranged from 2.4° to 7.6° and 0.11 mm to 0.39 mm, respectively. The differences in kinematics of lumbar segments between each combination of two sets of TMFs in 97% of cases for angular deformation and 55% of cases for translational deformation were within the reported accuracy of current imaging techniques. Therefore, it might be possible to use image-based kinematics of lumbar segments along with computational modeling for personalized assessment of TMFs.

1. Introduction

Neuromuscular control of spinal equilibrium and stability changes in the presence of pain or following exposure to known risk factors for low back pain (LBP) (Muslim et al., 2013; Radebold et al., 2000, 2001; Toosizadeh et al., 2013). Such alterations may cause deformations and/or forces in lower back tissues such that exceed injury/pain thresholds instantaneously or cumulatively (Adams et al., 2013; Coenen et al., 2014; Marras et al., 2001; Panjabi, 1992a,b). Despite such a significant role, the current methods for personalized assessment of trunk muscle forces (TMFs) are limited. Kinematic measures of lumbo-pelvic coordination, though capable of distinguishing patients with LBP from controls (Vazirian et al., 2016), do not provide much information about individual muscle forces. Specifically, neuromuscular redundancy in control of lumbo-pelvic motion as well as individual variability in mechanical behavior of passive lumbar tissues hinder relating measured kinematics data to TMFs. The commonly used surface electromyography (EMG)-based methods for the assessment of TMFs, on the other hand, can only provide information about the activity of superficial trunk muscles. Further, the relationship between EMG measures of muscle activity and actual muscle force is still unclear (Staudenmann et al., 2010). Finite element and multi-joint biomechanical models of the spine with detailed musculature have also been developed and used for general assessment of TMFs (Arjmand and Shirazi-Adl, 2006a,b; Dreischarf et al., 2014; Ezquerro et al., 2004; Hughes, 2000; Stokes and Gardner-Morse, 2001). These models often implemented optimization procedures to estimate TMFs (Arjmand and Shirazi-Adl, 2006b; Daniel, 2011; Hughes, 2000; Stokes and Gardner-Morse, 2001) and are not suitable for personalized assessment of TMFs due to assumptions made related to lumbar segmental rotations and the requirement for a priori knowledge of trunk neuromuscular strategy (e.g., a strategy that minimizes stress in muscles).

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Currently, imaging is used to detect structural and geometrical/kinematics abnormalities in the lumbar spine (Fujii et al., 2007; Iwata et al., 2013; Keller et al., 2003; Kjaer et al., 2005; Ochia et al., 2006). The image-based geometrical/kinematics information have also been used for development of geometrically personalized biomechanical models of normal and scoliotic spine (Eskandari et al., 2017; Ghezelbash et al., 2016; Lafon et al., 2010; Petit et al., 2004), biomechanical comparison of healthy and metastatically involved vertebrae (O'Reilly and Whyne, 2008), material sensitivity analysis of intervertebral disc (Fagan et al., 2002), indirect estimation of spinal loads (Shymon et al., 2014), and estimation of elastic modulus of cancellous bone (Diamant et al., 2005). The geometrical information from imaging if combined with optimization-based biomechanical models may provide a unique platform for personalized assessment of TMFs. Particularly, it will be possible to use an optimization-based biomechanical model to search for a set of muscle forces that results in lumbar kinematics similar to those obtained from imaging. Such a method, however, is reliable only if differences in lumbar spine kinematics due to differences in TMFs can be captured by the current imaging techniques.

Recently, we have used our finite element model of the spine within an optimization procedure to estimate TMFs and kinematics of lumbar segments resulting from a trunk neuromuscular strategy that minimized sum of squared stress across all trunk muscles (Shojaei et al., 2015). The resultant kinematics were consistent with image-based reports of lumbar spine kinematics of asymptomatic individuals. Using the proposed algorithm, estimation of TMFs and lumbar segmental kinematics for other hypothetical trunk neuromuscular strategies that optimize other aspects of lower back biomechanics is possible. As a first step toward testing the feasibility of using image-based kinematics of lumbar segments for personalized assessment of TMFs, therefore, the objectives of this short communication are to determine changes in lumbar segmental kinematics due to alterations in trunk neuromuscular strategy and the associated TMFs and to verify if such changes are within the reported precision of current imaging techniques.

2. Methods

To address our research questions, TMFs and lumbar segmental kinematics were estimated for five different trunk neuromuscular strategies. In our approach each neuromuscular strategy was represented by a distinct cost function for the optimization procedure and assumed to either represent the trunk neuromuscular strategy of asymptomatic persons or a neuromuscular abnormality that minimizes loading on a specific aspect of lower back tissues (i.e., muscles, ligaments, intervertebral discs, and facet joints). As noted earlier, a neuromuscular strategy associated with the minimum value of sum of squared muscle stresses across the entire trunk muscles resulted in lumbar segmental kinematics consistent with image-based reports of lumbar spine kinematics of asymptomatic individuals, hence, was regarded to represent a normal trunk neuromuscular strategy (Shojaei et al., 2015). On the other hand, abnormal neuromuscular strategies that minimize loads in muscles, ligaments, intervertebral discs, and facet joints were represented by cost functions that respectively minimize sum of squared muscle forces across the entire trunk muscles, passive moment, compression, and shearing force at the L5-S1 intervertebral disc. For each neuromuscular strategy, the associated TMFs and lumbar segmental kinematics for a single static trunk flexed posture involving, respectively, 40° and 10° of thoracic and pelvic rotations (i.e., equal to a total lumbar flexion of 40–10° = 30°) in the sagittal plane were estimated using our kinematics-driven finite element approach. Specifically, the changes in distance between centers of two vertebrae of each motion segment (i.e., translational deformation) as well as changes in their relative angular orientations with respect to each other (i.e., angular deformation) with alterations in TMFs were considered as changes in lumbar segmental kinematics. Forward trunk bending is a common posture used for X-ray imaging of patients with LBP and the specific thoracic and pelvic rotations considered here are the same rotations we used in a recent study for validation of our method (Shojaei et al., 2015).

In our approach, rather than implementing a force-driven approach for estimation of lumbar segmental kinematics resulting from TMFs that are associated with a given neuromuscular strategy, we used our kinematics-driven methods. Such a methodological choice was mainly because of the lower computational cost of kinematics-driven approach. Specifically, the potential TMFs that are searched in the optimization procedure, where a kinematics-driven approach is used, readily satisfy spine equilibrium. Hence, the solution space that is searched by the optimization search engine is much smaller than the case when a force-driven approach is implemented. Therefore, in our approach, from all possible sets of lumbar segmental kinematics that can be distributed across lumbar vertebrae and generate the total 30° lumbar flexion, we will search (i.e., through optimization procedures) for a set of lumbar segmental rotations where the associated biomechanical outcomes from the kinematics-driven approach minimize the desired cost function. Such a methodological choice (i.e., kinematics- versus force-driven), however, does not affect the outcomes. In the following subsections, we first elaborate on the kinematics-driven approach that uses lumbar segmental kinematics to estimate TMFs and other biomechanical outcomes (e.g., the L5-S1 passive moment) and subsequently present the structure of the optimization algorithm that finds the lumbar segmental rotations that optimize its cost function (i.e., representing a given neuromuscular strategy).

2.1. Estimating trunk muscle forces using the kinematics-driven approach

A nonlinear finite element (FE) model of spine, developed in the ABAQUS software (Version 6.13, Dassault Systémes Simulia, Providence, RI), is used in the kinematics-driven approach to estimate the moment at each lumbar vertebra to be balanced by muscles attached to that same vertebra (Arjmand et al., 2009; Bazrgari et al., 2007). In the FE model of spine, the thoracic region and lumbar spine vertebrae are simulated by rigid elements and intervertebral discs are simulated by nonlinear flexible beam elements (Fig. 1). Inputs to the FE model include sagittal plane rotational boundary conditions at the T12 to the S1 spinal levels along with the ~50% of total body weight distributed across the entire spine (Arjmand and Shirazi-Adl, 2006b). A muscle architecture including 56 muscles attached to the spine from lumbar and thorax to pelvis is considered for estimation of TMFs required to balance moments at lumbar vertebrae. Since the attached muscles to each level (i.e., 10 muscles in each level from T12 to L4 and 6 muscles in the level L5) outnumbers the moment equilibrium equations, an optimization procedure, hereafter called force optimization procedure, is used to estimate muscle forces at each level as follows:

$$\begin{cases} Var\ \boldsymbol{F} \\ Cost\ function = g(\boldsymbol{F}) \\ \text{Minimize}\ (cost\ function) \\ \text{Subject to}\ \sum_{i=1}^{m} r_i \times F_i = M \end{cases} \tag{1}$$