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Short communication

A rigid thorax assumption affects model loading predictions at the upper but not lower lumbar levels

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ABSTRACT

A number of musculoskeletal models of the human spine have been used for predictions of lumbar and muscle forces. However, the predictive power of these models might be limited by a commonly made assumption; thoracic region is represented as a single lumped rigid body. This study hence aims to investigate the impact of such assumption on the predictions of spinal and muscle forces. A validated thoracolumbar spine model was used with a flexible thorax (T1–T12), a completely rigid one or rigid with thoracic posture updated at each analysis step. The simulations of isometric forward flexion up to 80°, with and without a 20 kg hand load, were performed, based on the previously measured kinematics. Depending on the simulated task, the rigid model predicted slightly or moderately lower compressive loading than the flexible one. The differences were relatively greater at the upper lumbar levels (average underestimation of 14% at the T12L1 for flexion tasks and of 18% for flexion tasks with hand load) as compared to the lower levels (3% and 8% at the L5S1 for unloaded and loaded tasks, respectively). The rigid model with updated thoracic posture predicted compressive forces similar to those of the rigid model. Predicted muscle forces were, however, very different between the three models. This study indicates that the lumbar spine models with a rigid thorax definition can be used for loading investigations at the lowermost spinal levels. For predictions of upper lumbar spine loading, using models with an articulated thorax is advised.

1. Introduction

A number of biomechanical multi-body models of the human spine have been developed and used for estimations of lumbar spine loading (Arjmand and Shirazi-Adl, 2006; Christophy et al., 2012; de Zee et al., 2007; Shirazi-Adl et al., 2005; Stokes and Gardner-Morse, 1995). The knowledge of the loading conditions provided by these models is invaluable for improved understanding of the pathobiomechanics of low back pain, assessment of injury risk, development of prevention and treatment strategies and designing spinal implants. Even though these models have become progressively more complex and therefore more accurately represent the real spine anatomy and biomechanics, they are not free of simplistic assumptions, whose influences on the model predictions remain to be investigated.

A commonly made assumption in the lumbar spine models is that the thoracic region can be represented as a single lumped rigid body. The justification for this simplification is the observation that the thoracic spinal column (T1–T12), surrounded by the ribcage, has a considerably lower range of motion (~17–26°) than the lumbar region (~50–65°) (Hajibozorgi and Arjmand, 2016; Hsu et al., 2008; Mannion et al., 2004; Tully and Stillman, 1997). Nevertheless, a recent study indicates that including even the relatively small contribution of the thorax to the overall spine movement can be essential for appropriate estimation of gravity moments and may in turn greatly affect the predictions of the lumbar loading (Hajibozorgi and Arjmand, 2016).

The present study hence aims to investigate the impact of the rigid thorax assumption on the lumbar loading and trunk muscle force predictions. It is hypothesized that this assumption leads to an underestimation of the lumbar loads, increasing with the higher range of simulated flexion, due to increased muscular effort required to support the degrees of freedom of the thoracic segments.

2. Methods

2.1. Thoracolumbar spine model

The thoracolumbar spine model, developed in the AnyBody^{TM} modeling system (AnyBody Technology, Aalborg, Denmark) and validated for symmetric forward

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flexion tasks (Ignasiak et al., 2016) was used for simulations of flexion tasks with and without hand load. The following enhancements were introduced to the model in order to improve its computational stability and robustness, necessary to allow simulation of tasks involving large trunk flexion angles (80°).

- An assumed mass distribution of the thoracic segments (segmental masses and positions of centers of mass) was replaced by a more realistic one based on an experimental study by (Pearsall et al., 1996).
- The intercostales muscles, previously modeled as passive elastic elements, were replaced by active muscle fascicles located dorsally, laterally and ventrally in the intercostal spaces, representing both internal and external intercostales. The attachment points and physiological cross-sectional areas were estimated based on available data in literature (Kim et al., 2014; Wilson et al., 2001; Winter, 2009) (see Supplementary material 1).

Updating centers of masses had an almost negligible effect on the model predictions but enhancement of the intercostales muscles definition increased compressive forces for the middle to lower thoracic levels at flexed postures. Nevertheless, the predictions of the enhanced model for the intradiscal pressure correlated well with in vivo experimental data (see Supplementary material 2) and model robustness was considerably improved, expanding its applicability to a wider variety of tasks (such as large range of flexion or load lifting).

2.2. *Model configurations*

For the purpose of this study, the thoracolumbar spine model was used with the following thorax configurations:

A) Flexible: thoracic cage is represented as a fully articulated flexible musculoskeletal system (Ignasiak et al., 2016).
B) Rigid with updated kinematics: all joints between thoracic vertebra, ribs and sternum are replaced by rigid constraints but the positions of individual bony segments during simulated tasks correspond to those in the fully flexible model (thoracic posture changes during task). In other words, the thorax is effectively modeled as a single rigid body, but its shape depends on the posture.
C) Rigid: all joints between thoracic vertebra, ribs and sternum are replaced by rigid constraints; the relative positions of individual thoracic segments remain unchanged throughout the simulation, the thoracic posture is fixed, based on initial erect posture (i.e., the thoracic orientation reflects the orientation of T12 segment, which is determined by the lumbar and pelvis flexion.) This means the thorax is represented by a single rigid body of the same shape at every body posture.

In all configurations, the passive properties of the mobile intervertebral joints are defined by linear functions of rotational and translational displacements, with stiffness coefficients for thoracic and lumbar regions, based on experimental studies (Bisschop et al., 2012; Markolf, 1970; Panjabi et al., 1976; Schmidt et al., 1998), as described in (Ignasiak et al., 2016). In the lumbar region, the paraspinal ligaments are modeled as one-dimensional spring elements with the properties based on literature (Chazal et al., 1985; Pintar et al., 1992).

Even though the configuration that most closely corresponds to the common lumbar spine models is the fully rigid one, the rigid model with updated kinematics represents exactly the same body parts configuration (gravity moments and muscles lines of action) as the flexible model, except for the rigid connections between the thoracic cage segments. This intermediate configuration allows discriminating whether changes between flexible and rigid model predictions are a result of constraining degrees of freedom at the thoracic joints (otherwise actuated by muscles) or due to slightly different body postures.

2.3. *Simulations*

Using the model in these three configurations, inverse dynamics simulations (Damsgaard et al., 2006) of isometric symmetric forward flexion up to 80° (Fig. 1), with and without a 20 kg hand load, were performed. Average kinematics of the simulated trunk flexion were prescribed into the model based on our previous in vivo measurements on 21 young (age: 27.0 ± 4.0 years) healthy volunteers performing full range flexion maneuver (Ignasiak et al., 2015). In this optoelectronic motion capture measurement, beside the standard full body marker set, reflective spine markers were applied over the spinous processes of every other thoracic vertebra and every single lumbar vertebra (List et al., 2013). Based on approximating sagittal spine curvature by 3rd order polynomial function, the segmental flexion angles of C7T3, T3T5, T5T7, T7T9, T9T11, T11L1, L1L2, L2L3, L3L4, L4L5, and L5Sacrum, as well as pelvis flexion were calculated. For thoracic segments comprising more than one spinal unit, the segmental flexion was assumed to be shared evenly by the constituting articulations (e.g. T3T4 flexion is equal to T4T5 flexion).

The arms were assumed to be parallel to the gravity line throughout the task execution; hence arms and hand load were simulated by applying forces in the gravity direction on the uppermost 6 thoracic joints. For the case of the fully rigid thorax, upper trunk inclination angle was equal to sum of lumbar and pelvic flexion angles, (based on the same measured motion pattern but neglecting thoracic spine flexibility).

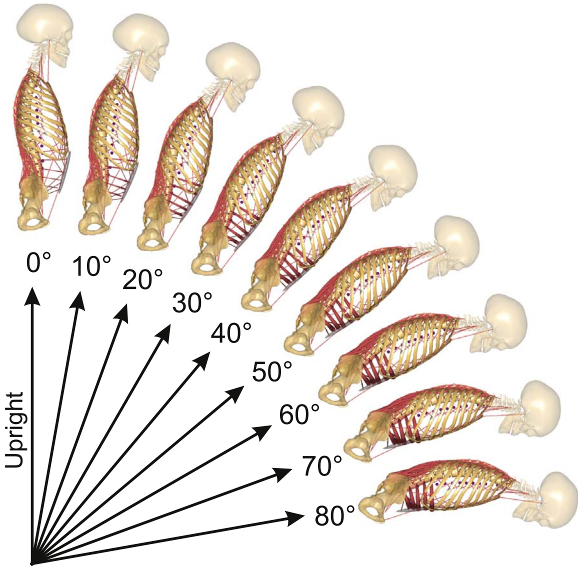

Fig. 1. Forward flexion task simulated with the rigid thorax model. The trunk inclination is determined by the orientation of the vector from the center of femoral head to T1 vertebra in the sagittal plane. (The transversus abdominis muscle is activated because it effectively acts as a spine extensor due to the intra-abdominal pressure definition; when the transversus forces act on the abdominal volume, forces extending the spine are applied on the lumbar vertebrae.).

3. Results

3.1. *Compressive loading predictions*

The differences in compressive forces predicted by the flexible thorax model, compared to fully rigid thorax model, varied from negligible to moderate, depending on the task, body posture and spinal level (Fig. 2). The rigid model predicted in general lower lumbar spine loading than the flexible one. The differences were relatively greater at the upper lumbar levels (i.e., T12L1 by 132 N or 14%, on average, for flexion task), especially for simulated flexion with hand load (347 N or 18%, on average), and decreased at lower levels (average underestimation at L5S1: 20 N or 3% and 131 N or 8%, for flexion and flexion with hand load, respectively). Updating the positions of the thoracic segments in the rigid model resulted in compressive forces predictions generally very similar to those of the fully rigid model.

3.2. *Shear forces predictions*

The shear predictions (computed in the mid-plane of the intervertebral disc) of the fully rigid model differed by −16 N to +131 N and those of the rigid model with updated kinematics by −19 N to +50 N, when compared to the flexible model for the simulated flexion tasks. For the 20 kg lifting tasks, these differences reached −63 to 175 N and −66 to 99 N, respectively. In general, the rigid model overpredicted the shear forces for the lumbar levels T12L1-L4L5, and slightly underpredicted them in the lowermost segment L5S1 (see Supplementary material 3 for more details).

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