Journal of Biomechanics 49 (2016) 959–966

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Thoracolumbar spine model with articulated ribcage for the prediction of dynamic spinal loading

Dominika Ignasiak [a,*], Sebastian Dendorfer [b], Stephen J. Ferguson [a]

[a] Institute for Biomechanics, ETH Zurich, Zurich, Switzerland
[b] Laboratory for Biomechanics, OTH Regensburg, Regensburg, Germany

ARTICLE INFO

Article history:
Accepted 10 October 2015

Keywords:
Thoracolumbar spine
Musculoskeletal model
Inverse dynamics

ABSTRACT

Musculoskeletal modeling offers an invaluable insight into the spine biomechanics. A better understanding of thoracic spine kinetics is essential for understanding disease processes and developing new prevention and treatment methods. Current models of the thoracic region are not designed for segmental load estimation, or do not include the complex construct of the ribcage, despite its potentially important role in load transmission. In this paper, we describe a numerical musculoskeletal model of the thoracolumbar spine with articulated ribcage, modeled as a system of individual vertebral segments, elastic elements and thoracic muscles, based on a previously established lumbar spine model and data from the literature. The inverse dynamics simulations of the model allow the prediction of spinal loading as well as costal joints kinetics and kinematics. The intradiscal pressure predicted by the model correlated well ($R^2=0.89$) with reported intradiscal pressure measurements, providing a first validation of the model. The inclusion of the ribcage did not affect segmental force predictions when the thoracic spine did not perform motion. During thoracic motion tasks, the ribcage had an important influence on the predicted compressive forces and muscle activation patterns. The compressive forces were reduced by up to 32%, or distributed more evenly between thoracic vertebrae, when compared to the predictions of the model without ribcage, for mild thoracic flexion and hyperextension tasks, respectively. The presented musculoskeletal model provides a tool for investigating thoracic spine loading and load sharing between vertebral column and ribcage during dynamic activities. Further validation for specific applications is still necessary.

1. Introduction

Musculoskeletal modeling offers an invaluable insight into the biomechanics of the healthy and pathological spine. A better understanding of the specific kinetics of the thoracic spine could cast light on the etiology of spinal disorders typical for this region, such as hyperkyphosis (Bruno et al., 2012; Finocchiaro et al., 2012; Huang et al., 2006; Katzman et al., 2010; Sinaki et al., 2005), osteoporotic and traumatic fractures (De Smet et al., 1988; Johnell and Kanis, 2006; Melton III et al., 1999; Rao et al., 2014; Vialle and Vialle, 2005; Wood et al., 2014), scoliosis (Grauers et al., 2014; Gummerson and Millner, 2011; Hebela and Tortolani, 2009; Nnadi and Fairbank, 2010; Weinstein et al., 2008), or upper back pain (Bradshaw et al., 2011; Cho et al., 2012; Niemelainen et al., 2006), leading to improved treatment and prevention planning.

Previous modeling works dedicated to the thoracic region have focused on ribcage deformities (Aubin et al., 1995; Closkey et al., 1992; Roberts and Chen, 1970), passive stability provided by the ribcage to the spine (Andriacchi et al., 1974; Sham et al., 2005), respiratory mechanics (Loring and Woodbridge, 1991; Ratnovsky and Elad, 2005), vibration response of the spine (Kong and Goel, 2003) or corrective treatments of scoliosis (Aubin et al., 2003; Duke et al., 2005; Grealou et al., 2002; Perie et al., 2003; Salmingo et al., 2012; Wang et al., 2014), but did not explore loading patterns of this region during daily living activities. On the other hand, most of the spine models developed for load estimation have been devoted to the lumbar region, with the thorax represented as a single rigid body (Arjmand et al., 2009; de Zee et al., 2007; Galbusera et al., 2013; Han et al., 2012; Shirazi-Adl et al., 2005; Stokes and Gardner-Morse, 1995), thus not capable of predicting forces between its components. A few thoracolumbar models with discrete thoracic vertebral bodies either completely discarded the contribution of the ribcage (Briggs et al., 2007, 2006; Keller et al., 2005) or approximated its effects from the stiffness

* Correspondence to: Institute for Biomechanics, ETH Zurich, Hönggerbergring 64, HPP-O24, 8093 Zurich, Switzerland. Tel.: +41 44 633 79 51; fax: +41 44 633 11 24.
E-mail address: dignasiak@ethz.ch (D. Ignasiak).

http://dx.doi.org/10.1016/j.jbiomech.2015.10.010
0021-9290/

properties (Iyer et al., 2010) without modeling explicitly its individual structures and their mechanical interactions. However, beside the reported stiffening effect of the ribcage on the thoracic spine (Brasiliense et al., 2011; Sham et al., 2005; Watkins et al., 2005), several studies and clinical observations suggest that its components play an important role in force transmission (Berg, 1993; Fowler, 1957; Klaase et al., 1998; Lund et al., 2001; Muldoon et al., 1999).

Therefore, the main aim of this work was to develop and validate a generic detailed model of the entire thoracolumbar spine with articulated ribcage, which can be used to estimate load distribution between components of the spinal column and ribcage during dynamic tasks. We hypothesized that explicit modeling of ribcage components, when compared to simulating only their stiffening effect on vertebral joints, will result in a reduction or broader distribution of the predicted thoracic segmental loading.

2. Methods

The model was constructed with the AnyBody Modeling System (AMS, v. 6.0), which is commercial software (AnyBody, Aalborg, Denmark) for three-dimensional multi-body dynamic simulation (Damsgaard et al., 2006). A previously established detailed generic lumbar spine model (de Zee et al., 2007) [presented in Appendix 1], was incorporated as a basis for the development of the thoracic part.

2.1. Thoracic spine model – modifications to the lumbar spine model

2.1.1. New segments of the thoracic region

Individual rigid bodies were defined within the thoracic region to represent the 12 thoracic vertebrae, bony fragments of the 10 pairs of ribs, and a sternum, resulting in 33 new segments. The total thoracic mass defined in the lumbar spine model (de Zee et al., 2007), as based on (Winter, 2009), was distributed between thoracic vertebrae, ribs and sternum. The respective positions of costal and vertebral centers of mass (Fig. 1) were defined so that the overall resultant thoracic center of mass was preserved, as in the previously estimated location.

2.1.2. Joints

The intervertebral joints in the thoracic spine were defined as 6 degrees-of-freedom joints, allowing rotation and translation in all directions. The same joint definition was used to model joints between the lumbar vertebrae, adding translational degrees-of-freedom to the previously spherical joints. The articulations of the ribs with the vertebral bodies, namely costo-vertebral and costo-transverse joints, were modeled as single compound revolute joints (Minotti and Lexcellent, 1991) with the rotation axis in the frontal direction (Fig. 2A–C). The articulations of the ribs through deformable costal cartilages and synovial joints to the sternum were modeled as 6 degrees-of-freedom joints for all ribs, except the first pair, which was modeled as spherical joint, since the first pair of ribs articulates with the sternum via a less-mobile cartilaginous junction (Fig. 2D).

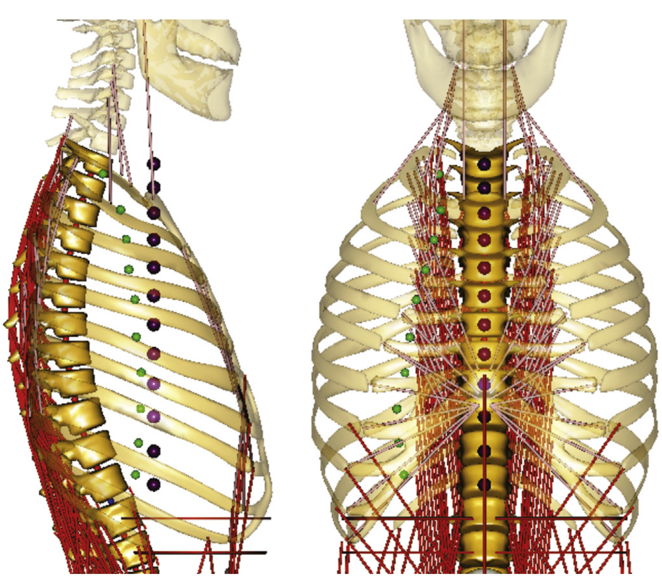

Fig. 1. The centers of mass of the thoracic vertebrae (purple) and right ribs (green), lateral and anterior view. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

2.1.3. Muscles of the thoracic spine and ribcage

The attachments of muscles spanning from the lumbar to the thoracic region, already present in the base lumbar spine model, were defined in the corresponding reference frames of the thoracic vertebrae, ribs or sternum. In order to control degrees-of-freedom of the newly defined segments, in total 227 fascicles of thoracic long and short muscles, as well as ribcage muscles [for details see: Appendix 2], were introduced (Fig. 3) based on anatomical descriptions (Bakkum and Cramer, 2014). Muscles were modeled as active force elements passing over several bony structures, which transfer the force to origin and insertion points in a longitudinal direction along the muscle path, and at via-nodes in the direction of a line bisecting the angle formed by the muscle path on both sides of a node.

2.1.4. Elastic properties

The rotational stiffness of the lumbar vertebral segments, mimicking the combined effects of the intervertebral disks, paraspinal ligaments and facet joints, modeled as a linear function of angular displacement in three directions (Schmidt et al., 1998), was incorporated from the basic lumbar spine model (de Zee et al., 2007). The rotational stiffness of the thoracic vertebral segments was defined using values from experimental studies (Markolf, 1970; Panjabi et al., 1976). In the model without the ribcage, they were increased (White and Panjabi, 1990) in order to represent the ribcage by its stiffening effect on the spine. Translational stiffness forces, representing the compressive and shear deformations of the lumbar and thoracic spinal segments, were applied proportionally to the linear displacements of the vertebral bodies, based on values reported in (Bisschop et al., 2012; Markolf, 1970; Panjabi et al., 1976), which along with rotational stiffness coefficients are summarized in Table 1.

The total stiffness of the costo-vertebral articulations, resulting from the joint capsules and the surrounding ligaments, was modeled with two-directional linear and three-dimensional rotational elastic elements between the rib and lower vertebral body. The overall stiffness of costo-transverse joints, their capsules and ligaments, was represented by elastic elements between the ribs heads and transverse processes. The properties of the costal cartilages and costo-sternal joints were simulated by elastic bands between the ribs and sternum. The intercostal muscles between the ribs were modeled as passive one-dimensional elastic elements. The stiffness coefficients of the elastic properties of the ribcage (Table 2) were taken from the simulation estimates (Andriacchi et al., 1974), found to match the cadaveric force–displacement measurements (Schultz et al., 1974).

2.2. External loads

In the current model the pelvis is constrained to the ground. The load and moment related to the neck and head are applied on the first thoracic vertebra (T1). The weight of the arms and (optionally) carried load, as well as approximated moment related to the carried load, is distributed between the six upper thoracic joints (T1T2–T6T7).

2.3. Calculation of muscle forces and spinal loads

In order to predict muscle forces and joint reaction forces during simulated tasks, a combination of a traditional inverse dynamics approach and a more recently developed algorithm for the analysis of small motions in individual joints has been employed in this study. By solving numerically the Newtonian equations of motion at every time step, the internal forces necessary to perform a defined motion are calculated. The redundancy problem of this method is overcome by employing muscle recruitment optimization strategies based on physiological criteria and described in detail by (Damsgaard et al., 2006; Rasmussen et al., 2001); in this work a third-order polynomial objective function was chosen for minimizing muscle activation, thus fatigue, while maximizing muscle synergy.

In order to predict the effect of the small displacements that otherwise would be very difficult to define, as translational displacements of the vertebrae related to compressive and shear deformations of the intervertebral discs, and rotational and translational motion of the ribs with respect to the vertebrae and the sternum, the force dependent kinematics (FDK) method (Andersen et al., 2011) has been used. It simultaneously computes muscle, joint and ligament forces as well as internal joint kinematics by combining inverse dynamics and quasi-static force equilibrium in selected degrees of freedom. Computationally, additional kinematic constraints are added to the inverse dynamics analysis to obtain a kinematically determined system. From this, supplementary forces necessary to balance the system are computed and the joint position is updated according to the direction of the forces. The supplementary forces are then minimized through iterations between the inverse dynamics and static equilibrium analyses until convergence is reached. Thus, an FDK subroutine for each global inverse dynamics step is included.