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Sensitivity of lumbar spine loading to anatomical parameters

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ABSTRACT

Musculoskeletal simulations of lumbar spine loading rely on a geometrical representation of the anatomy. However, this data has an inherent inaccuracy. This study evaluates the influence of defined geometrical parameters on lumbar spine loading utilising five parametrised musculoskeletal lumbar spine models for four different postures. The influence of the dimensions of vertebral body, disc, posterior parts of the vertebrae as well as the curvature of the lumbar spine was studied. Additionally, simulations with combinations of selected parameters were conducted. Changes in L4/L5 resultant joint force were used as outcome variable. Variations of the vertebral body height, disc height, transverse process width and the curvature of the lumbar spine were the most influential.

These parameters can be easily acquired from X-rays and should be used to morph a musculoskeletal lumbar spine model for subject-specific approaches with respect to bone geometry. Furthermore, the model was very sensitive to uncommon configurations and therefore, it is advised that stiffness properties of discs and ligaments should be individualised.

1. Introduction

Knowledge about lumbar spine biomechanics is important to comprehend the emergence of low back pain and to develop the optimal clinical treatment methods as well as new spinal implants. Over the last decades, computer simulation proved a powerful tool to acquire this knowledge. However, the high level of variability in biological systems complicates the choice of the best suited model. Especially, the examination of patient-specific cases requires individualised models for accurate and reliable computations. Even though it is rather straight forward to acquire 3D bone data from the use of state-of-the-art segmentation software on clinical image data, the process remains a time consuming and non-automated procedure. Also, it is uncertain what level of accuracy is needed regarding the patient-specific bone geometry and a single parametric variation may often not be sufficient to cover the entire inter-subject variability. Regarding this, Cook et al. (2014) reviewed literature and highly advocated a broader treatment of the variability by utilising more sensitivity analysis.

For example, Niemeyer et al. (2012) studied a fully parameterised and geometrically simplified FE model to supply information about the influence of the natural variability on biomechanics and they reported the disc geometry and the facets position as important. FE studies revealed further important facts concerning the lumbar spine, e.g. Dreischarf et al. (2013) analysed relations between compressive force and intradiscal pressure (IDP) and reported that in vivo compression forces can be estimated within a certain range from IDP values.

Those FE studies are restricted to a part of the spine and while they show limitations regarding the load application of muscle forces they incorporate highly detailed material properties. In contrast to that, musculoskeletal models often consist of a full human body and utilise rigid bodies for bones and different mechanical material models for muscles or ligaments. The spine model published by de Zee et al. (2007) is frequently used as a basis for various simulations because it includes a detailed lumbar spine and it is possible to simulate a variety of postures. For instance, Han expanded the model with regard to muscles, ligaments, disc stiffness and intraabdominal pressure and studied the influence of increasing body height and weight on lumbar segmental loads where a linear influence of the parameters was reported (Han et al., 2012b, 2012a). Others used outputs of the model to investigate lumbar muscle activity, kinematics and

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kinetics during slipping (Rashedi et al., 2012) or examined the effects of posterior lumbar spine surgery on muscle activity (Bresnahan et al., 2010).

However, to the authors' knowledge the influence of specific dimensions of lumbar vertebrae on loading in the lumbar spine has not been studied systematically. The goal of this study was to identify specific lumbar dimensions with large influences on lumbar spine loading.

2. Material and methods

2.1. Test subjects

Clinical image data of five male test subjects were used to generate different lumbar models. Table 1 lists the subjects' data. All subjects showed age-related degenerations of the lumbar spine but none had undergone a surgery. Furthermore, subject one was diagnosed with a degenerative disc at the L5/S1 level and beginning degenerations in the other four lumbar discs while subject four exhibited a severe degeneration of the disc at the L4/L5 level. Comparison of the anthropometric data of the subjects with published anatomical studies showed that they distribute well over the reported vertebral sizes (Berry et al., 1987; Panjabi et al., 1992, 1993; Scoles et al., 1988).

2.2. Musculoskeletal model

The AnyBody Modeling System (AMS, Version 6.0.3.4196) was used for the musculoskeletal simulations and the AnyBody Managed Model Repository (AMMR, Version 1.4.1) provided different basic body parts. The full body model used in this study included the following: both legs and arms, the skull and a detailed customised spine model, whereas legs, arms, skull and upper thorax were taken from the AMMR. The model included the thoracic and cervical spine as well as the rib cage as one lumped part. Adjacent lumbar vertebrae were connected via a spherical joint including a three-dimensional rotational stiffness with a linear relationship (Schmidt et al., 1998) representing the disc stiffness. Muscles were modelled as active elements with a constant strength. Their line of action is defined by origin, insertion and via-points. The muscle force is transferred along this path and the via-points enable muscles to transfer a force to the connected segments in the direction of a line which bisects the angle formed by the muscle path. Ligament properties were derived from literature (Chazal et al., 1985; Pintar et al., 1992). This basic model is described in detail by de Zee et al. (2007).

The lumbar vertebral geometries were segmented from clinical image data. The 3D models were used to morph the nodes representing muscle and ligament attachment points of the base model to the subject-specific positions. All other segments were scaled with the height and weight information according to the scaling law covered by Rasmussen (2005). Ten linear elastic elements (tension only) at each lumbar level represented the ligaments with the ligamentum flavum modelled with two elements and the intertransverse ligament with four elements. The individual geometry of adjacent facets was used in the surface contact simulation. The contact model is based on penetration volume and a stiffness factor, which was tuned to replicate the relatively hard bone/cartilage surface without terminating the simulation. Moreover, the location of centres of rotation (CoR) in the sagittal plane between motion segments from L1 to L5 was dependent on vertebral body size according to mean positions for a flexion movement from literature data (Pearcy and Bogduk, 1988). The position of the CoR in mediolateral direction was set to the central plane of the inferior vertebral body. Furthermore, lumbar lordosis in a standing posture was adjusted according to the clinical image data. The final model consisted of 27 rigid bodies for various bones, 287 muscles and 60 linear elastic elements representing the lumbar ligaments.

The collective centre of mass (CoM) of the model was computed by the AMS based on mass and position of every single segment. Constraints fixed the CoM above the ankles in the anterior–posterior direction and centred it between the feet while flexion in the ankle joint remained unconstrained. Furthermore, slack lengths of all lumbar ligaments were adjusted to the upright standing posture and therefore their tensile force was set to zero in this position. In addition, all rotational movements between the lumbar segments were computed by the AMS. This solver option, called force-dependent kinematics (FDK) in AMS terms, locally suspends the inverse dynamic approach and uses forward dynamics to determine the rotational positions. More precisely, a quasi-static force equilibrium is assumed balancing the acting forces and moments at the motion segments in an iterative process where the rotation in the joint is used as the variable. The forces and moments are generated from muscles, ligaments, disc stiffness as well as the reaction forces in the joint (Andersen et al., 2011). The criterion for a successful iteration was set to a remaining force residual of less than 0.5 N. If this criterion could not have been met after a set maximum number of iterations (400) the simulation was terminated and marked as failed.

In order to solve the redundancy of muscles of the musculoskeletal system an optimisation is implemented in the AMS (Damsgaard et al., 2006). The according criterion is represented by Eq. (1) which represents a polynomial muscle recruitment criterion with the power of 3. This equation is minimised and considers the muscle forces f_i and a normalisation factor N_i. In this study the cross sectional area of a muscle was used as normalisation factor.

$$G = \sum \left(\frac{f_i}{N_i} \right)^3 \tag{1}$$

Table 1
Data of test subjects.

Model number	Sex	Age	Weight (kg)	Height (cm)	Lordosis angle in deg (Cobb's method)
1	m	unknown	70	168	60
2	m	64	86	177	53
3	m	58	97	178	43
4	m	55	124	174	57
5	m	52	89	185	54

2.3. Validation of IDP at the L4/L5 level

The models were validated against experimental data of IDP measurements at the L4/L5 level (Wilke et al., 1999; Sato et al., 1999; Takahashi et al., 2006) and one at the L3 level (Nachemson, 1975). This data was processed to compare relative changes of IDP between a standing posture and various daily activities. Hence, various postures described in the publications were simulated with every model, L4/L5 resultant joint forces were then referenced to the results from the standing posture and compared to the changes of the experimental data. Moreover, the task which consolidated flexion and lifting a weight was studied in an additional simulation with different lumbar flexion angles (ranging between 10° and 50°).

Furthermore, the absolute values of the computed IDP between L4/L5 in an upright standing posture were compared to literature data. Therefore, the computed compressive forces were converted to an IDP via Eq. (2) which uses the subjects' disc cross-sectional areas (A_{disc}), the compressive force in a standing posture and a correction factor (CF). The disc areas were measured from the clinical images. Regarding the CF, Nachemson (1960) reported a mean *CF* of 0.66 and according to Dreischarf et al. (2013) various in vitro studies indicated a range of 0.55–0.77 for an individual CF.

$$IDP = \frac{F_{compressive}}{A_{disc} \cdot CF_{min/mean/max}} \tag{2}$$

Additionally, the compressive force between L4/L5 was computed for the following basic motions: lateral bending from left to right, an extension to flexion movement and torsion. Afterwards, the results were converted to IDP values and compared to literature (Wilke et al., 2001).

2.4. Parameter study

Fig. 1 shows an overview of the analysed postures and parameters. In the first set of simulations, only one parameter was changed to 11 different sizes. The alteration affected all lumbar levels equally and changed the location of attachment points of muscles and ligaments. Afterwards, additional simulations were carried out which addressed a set of combinations of the parameters showing a larger influence on lumbar loading in the previously performed simulations. With five different models and each in four different postures, the studies with a single parameter totalled 2200 simulations while the studies with simultaneous parametric changes totalled 1500 simulations. All in all 3700 computations were conducted. Table 2 lists all simulations per posture with their according variables and intervals. The baseline values of every interval were the sizes of the subjects' vertebrae. The interval sizes corresponded to the largest standard deviations of the dimension derived from literature data (Panjabi et al., 1992, 1993; Scoles et al., 1988). The interval for the lordosis angle was set to a value which allowed a reasonable positioning of the lumbar vertebrae.

The analysed static postures comprised upright standing, 50° flexion, 10° axial rotation and 15° lateral bending. Those postures were measured between the pelvis segment and the thorax segment.

The resultant joint force between L4 and L5, computed as reaction force in the spherical joint, was used as outcome parameter. Evaluation of the results was conducted in Matlab (Version R2014a).

3. Results

The following section presents the results obtained for model validation as well as the data acquired from the parameter studies.