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Technical note

Optimised *in vitro* applicable loads for the simulation of lateral bending in the lumbar spine

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ABSTRACT

In *in vitro* studies of the lumbar spine simplified loading modes (compressive follower force, pure moment) are usually employed to simulate the standard load cases flexion-extension, axial rotation and lateral bending of the upper body. However, the magnitudes of these loads vary widely in the literature. Thus the results of current studies may lead to unrealistic values and are hardly comparable. It is still unknown which load magnitudes lead to a realistic simulation of maximum lateral bending.

A validated finite element model of the lumbar spine was used in an optimisation study to determine which magnitudes of the compressive follower force and bending moment deliver results that fit best with averaged *in vivo* data.

The best agreement with averaged *in vivo* measured data was found for a compressive follower force of 700 N and a lateral bending moment of 7.8 N m.

These results show that loading modes that differ strongly from the optimised one may not realistically simulate maximum lateral bending. The simplified but *in vitro* applicable loading cannot perfectly mimic the *in vivo* situation. However, the optimised magnitudes are those which agree best with averaged *in vivo* measured data. Its consequent application would lead to a better comparability of different investigations.

1. Introduction

In *in vitro* experiments and finite element calculations of the lumbar spine simplified loading modes in the three main anatomical planes are usually employed to simulate upper body movements [1]. Recommendations for realistic simulation when using those simplified loading modes exist for axial rotation [2] and flexion-extension [3] and are, *e.g.* required for realistic pre-clinical tests of new spinal implants. However, no recommendation has yet been given for lateral bending. Lateral bending is often simulated with the help of two *in vitro* applicable loading techniques: a compressive follower force and a pure bending moment. Moments between 3.75 and 18 N m [4,5] and a compressive follower force between 0 and 1200 N [5,6] have been used in published studies. Due to these varying load magnitudes, the results of current investigations may take on unrealistic values and are hardly comparable. It is still unknown which load magnitudes lead to a realistic simulation of maximum volunteer lateral bending of the spine.

Only a few mechanical parameters such as intradiscal pressure (IDP) and intervertebral rotation (IVR) have been measured in the spine *in vivo* and can be used to characterise the *in vivo* spinal load and motion during lateral bending of the upper body. IDP in volunteers during lateral bending while standing were measured by Nachemson [7] and Wilke et al. [8], who found a maximum increase of the IDP on average of 36% and 20%, respectively, in comparison with that of relaxed standing. *In vivo* measurements of the spinal kinematics for the same motion were performed by Plamondon et al. [9] and Pearcy and Tibrewal [10], using radiographic techniques and by Steffen et al. [11,12], who used Kirschner wires implanted into the spinous processes and electromagnetic tracking sensors. They showed that segmental IVRs have the value of about 5.1° (mean calculated for levels between L1 and L5) in the lumbar spinal segments during lateral bending (Table 1).

The aim of this study was to optimise the magnitudes of the compressive follower force and bending moment to find an *in vitro* applicable loading mode for lateral bending where the calculated parameters IVR and IDP agree best with averaged values taken from

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Table 1
In vivo measured segmental intervertebral rotations for lateral bending of different studies and their averages.

Study	Number of subjects	Segment				
		L1–L2	L2–L3	L3–L4	L4–L5	L5–S1
Pearcy and Tibrewal [10]	10	5.5°	5.5°	5.0°	2.5°	−1.0°
Plamondon et al. [9]	4	4.9°	5.7°	4.5°	4.9°	–
Steffen et al. [11,12]	16; 8	–	–	6.3°	5.1°	–
Averaged *in vivo* data	–	5.2°	5.6°	5.3°	4.2°	–

in vivo measured data. For the optimisation procedure, a validated finite element model of the lumbar spine was used.

2. Methods

2.1. Finite element model of the intact lumbar spine

A validated, non-linear finite element model of the osteoligamentous lumbar spine was used in this optimisation study (Fig. 1). The model and its extensive validation for IVR and IDP were previously reported in detail [13–15]. In brief, the symmetrical model consists of five vertebrae, five intervertebral discs and all spinal ligaments. The nuclei pulposi were simulated as incompressible fluid-filled cavities which were surrounded by the annuli fibrosi modelled as fibre-reinforced hyperelastic composites. The fibres of the annuli were embedded in the ground substance in concentric rings around the nuclei in 14 layers, arranged in a criss-cross pattern. The vertebrae possess curved facet joints with a thin cartilage layer which was simulated using soft contact with exponentially increasing contact force for decreasing contact gap [16]. When unloaded, these joints have a gap of 0.5 mm. Material properties of the different tissues were taken from the literature (Table 2).

The numerical calculations were performed using the finite element program ABAQUS, version 6.10 (Simulia Inc., Providence, RI, USA). Segmental intervertebral rotations during lateral bending were described using Cardan angles (sequence: flexion-extension, lateral bending, and axial rotation). The component for lateral bending of these Cardan angles was used to characterise the relative vertebral lateral bending between two adjacent vertebrae.

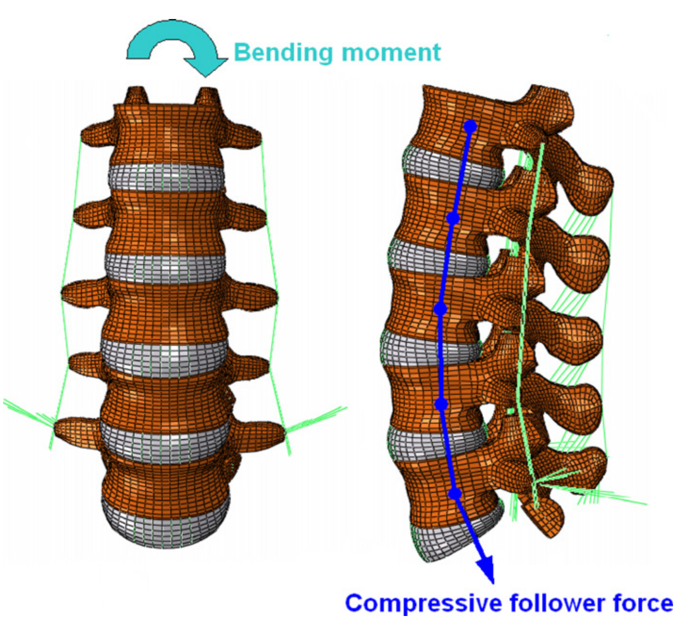

Fig. 1. Employed finite element model of the lumbar spine and loads applied in the optimisation study.

In a first loading step, a compressive follower force [17–19], that was optimised in a previous study and caused less than 0.25° IVR in the sagittal plane and no IVR in the frontal plane, was applied in the median sagittal plane of each vertebra. In a second loading step, a lateral bending of the lumbar spine was simulated by applying an additional pure moment in the frontal plane at L1. In all loading steps, the most cranial vertebra L1 was unconstrained in accordance to Wilke et al. [1], thus allowing coupled motion as it exists *in vivo* [10].

2.2. In vivo reference values for lateral bending

In vivo measurements of IDP and IVR are highly complex and have been performed mostly only in a few subjects at selected levels. Therefore, the results of different studies were averaged in order to increase the relevance of the single IVR (Table 1) and IDP values. Only similar studies were included where lateral bending was actively performed from an upright standing position with maximum voluntary effort in healthy people. Furthermore, IVR were only considered when more than one *in vivo* measurement had been performed (Table 1). Thus, the level L5–S1 was not considered in the optimisation study. The IDP measurements by Wilke et al. (disc L4–L5) [8] and Nachemson (disc L3–L4) [7] were also averaged, which led to a mean increase of the IDP of 28% in lateral bending in comparison to upright standing. However, the absolute value of the IDP in relaxed standing was taken from measurements of Wilke et al. [8]. They measured an IDP of 0.5 MPa at level L4–L5 during relaxed standing, using advanced techniques in a healthy volunteer with proven non-degenerated intervertebral discs. For lateral bending, this procedure leads to an averaged IDP of 0.64 MPa.

2.3. Optimisation study and subsequent robustness analysis

The optimisation of the magnitude of the compressive force and bending moment was performed using a gradient-based optimisation algorithm (optiSLang, Dynardo GmbH, Weimar, Germany). The software uses the NLPQLP (non-linear programming using a quadratic or linear least-square algorithm parallel line search) algorithm [20]. This algorithm adjusts the amounts of the compressive force and moment by minimising the following objective function (f_{OBJ}):

$$f_{OBJ} = \left(\frac{IVR_{L1-L2} - 5.2^\circ}{5.2^\circ}\right)^2 + \left(\frac{IVR_{L2-L3} - 5.6^\circ}{5.6^\circ}\right)^2 + \left(\frac{IVR_{L3-L4} - 5.3^\circ}{5.3^\circ}\right)^2 + \left(\frac{IVR_{L4-L5} - 4.2^\circ}{4.2^\circ}\right)^2 + \left(\frac{IDP_{L4-L5} - 0.64\,\text{MPa}}{0.64\,\text{MPa}}\right)^2$$

The objective function contains the relative deviation between the calculated values for the segmental IVRs (L1–L5) and the IDP (L4–L5), and the averaged *in vivo* values for IVR (Table 1) and IDP (0.64 MPa).