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# Computational biomechanics of a lumbar motion segment in pure and combined shear loads

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ABSTRACT

Anterior shear has been implicated as a risk factor in spinal injuries. A 3D nonlinear poroelastic finite element model study of a lumbar motion segment L4-L5 was performed to predict the temporal shear response under various single and combined shear loads. Effects of nucleotomy and facetectomy as well as changes in the posture and facet gap distance were analyzed as well.

Comparison of the predicted anterior displacement and stiffness response with available measurements indicates satisfactory agreement. Under shear loads up to 400 N, the model predicted an almost linear displacement response. With increasing shear load and/or compressive preload, the stiffening behavior becomes evident, primarily due to stretched collagen fibers and greater facet interactions. Removal of the facets markedly decreases the segmental stiffness in shear and thus highlights the importance of the facets in resisting shear force; 61–87% of the applied shear force is transmitted through the facets depending on the magnitude of the applied shear and compressive preload. Fluid exudation during the day as well as reduced facet gap distance and a more extended posture yielded higher facet joint forces. The shear resistance of the motion segment remains almost the same with time despite the transfer of load sharing from the disc to facets.

Large forces on facet joints are computed especially under greater compression preloads, shear forces and extension rotations, as time progresses and with smaller gap distances. The disc contribution on the other hand increases under larger shear loads, smaller compression preloads, flexed postures, larger facet gap distances and at transient periods.



## 1. Introduction

During various daily activities, the human lumbar motion segments are subjected to a wide range of loading and displacements. Investigations of the motion segment response under static and transient loads in compression, flexion, extension, lateral bending and torsion have been the subject of many experimental and theoretical studies. However, despite the ever presence of shear forces at all spinal levels under occupational and recreational activities, only few works have studied the behavior of the lumbar spine in shear. The shear–flexion ratios have been related to symptoms observed in patients with degenerative lumbar instability (Stokes and Frymoyer, 1987). Moreover, lumbar shear forces have been linked to an increase in the rate of back injuries (McGill et al., 2000; Norman et al., 1998).

In experimental studies, the contribution of individual structures of the motion segments in resisting applied anterior shear loads was investigated by comparison of the response with and without posterior elements (Lu et al., 2005; Skrzypiec et al., 2012). In these studies, the disc contributed only 23–34% of the applied shear load under small displacements. At higher displacements, the disc load-bearing increased to 66% at 5 mm and to 56% at failure (Skrzypiec et al., 2012). This highlights the importance of the facets and posterior ligaments in resisting shear loads. A finite element (FE) analysis indicated that the disc resists about 65% of the applied shear load of 150 N (Teo et al., 2004). Interestingly, no stiffening effect in the global response was observed due likely to the negligible mechanical role of collagen fibers under such small shear forces.

There exists however no studies on the effects of different compression preloads and coupled flexion/extension angular displacements (*i.e.*, posture) on the segmental shear stiffness as well as the load-bearing role of the disc and facet joints.

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Furthermore, the shear response under physiological diurnal compression preload has neither been investigated. Such studies are crucial in devising effective measures to prevent or reduce the risk of injury and to improve the treatment and rehabilitation of injured spines.

The objective of the present work was hence set to initially validate a poroelastic FE model of an intact L4-L5 lumbar motion segment and its isolated intervertebral disc under shear loading. Both models were subsequently employed to investigate the shear stiffness and load-bearing response under various loading conditions. It was hypothesized that

(1) in contrast to the clear stiffening effect reported under increasing compression, bending or torsion (Shirazi-Adl et al., 1986a, 1986b) and due to negligible role of disc fibers, no such stiffening effect appears under unconstrained shear loading,
(2) the shear stiffness increases under greater compression preloads. This hypothesis is based on previous *in vitro* and *in silico* studies showing that compression preloads substantially increase the disc stiffness under rotations/moments in different planes (Gardner-Morse and Stokes, 2003; Goel et al., 1993; Rohlmann et al., 2009b; Shirazi-Adl et al., 1986a, 1986b) and
(3) the shear resistance of the motion segment increases with time as fluid exudes from the disc under creep compression preload and as facets interactions increase.

## 2. Materials and methods

### 2.1. FE model

A previously published 3D poroelastic nonlinear FE model of the L4-L5 human lumbar spine segment was used (Schmidt et al., 2011) as the reference model (RM) for subsequent parametric investigations. In brief, the RM consists of nine distinct structural regions, namely cancellous bone, cortical bone, posterior bony elements including the facet articulations, inner and outer annulus, nucleus, cartilaginous and bony endplates plus seven major ligaments. All structures, except the posterior bony elements, were modeled as biphasic materials consisting of a solid phase fully saturated by an incompressible fluid phase.

The annulus was assumed as a composite of collagen fibers embedded in a matrix of ground substance. The articulating facet surfaces were modeled using surface-to-surface frictionless contacts. For the current study, the initial gap between the articulating surfaces was taken to be nil. The facet cartilage layers had a uniform thickness of 0.4 mm and were represented by a Mooney–Rivlin hyperelastic material ($c_{10}$=0.07, $c_{01}$=4.95, and $D$=0.45). After mesh convergence studies, the RM consists of 2706 20-node isoparametric solid elements and 924 membrane elements with a total of 41,696 degrees of freedom (Fig. 1).

### 2.2. Material properties

The drained solid phases of the vertebral body and endplates were modeled as linear elastic isotropic while the annulus bulk and the nucleus were represented as a compressible neo-Hookean hyperelastic material (Galbusera et al., 2011; Schmidt et al., 2013) (Table 1). The total stress ($\sigma$) in the tissue was expressed as

$$\sigma = -(\mu^f + \Delta\pi)I + \sigma_S \tag{1}$$

with $\mu^f$ being the water chemical potential, $I$ the identity matrix, $\sigma_S$ the effective stress within the solid skeleton and $\Delta\pi$ the osmotic pressure gradient calculated as (Wilson et al., 2005)

$$\Delta\pi = \varphi_i RT\sqrt{c_F^2 + 4\frac{\gamma_e^2}{\gamma_i^2}c_e^2} - 2\varphi_e RTc_e \tag{2}$$

where $\varphi_i$ and $\varphi_e$ are respectively the internal and external osmotic coefficients, $\gamma_i$ and $\gamma_e$ the activity coefficients (assumed equal herein), $c_e$ the salt concentration in the external bath, $R$ the universal gas constant, $T$ the absolute temperature and $c_F$ the fixed charge density per water content which depends on the tissue deformation

$$c_F = c_{F0}\left(\frac{n_0}{n_0 - 1 + J}\right) \tag{3}$$

with $c_{F0}$ being the initial fixed charge density, $J$ the deformation tensor and $n_0$ the initial fluid volume fraction.

Strain-dependent permeability $k$ was implemented as a function of the current void ratio $e$, which varies with tissue deformation, its initial value $e_0$, initial permeability $k_0$ and an empirical positive coefficient $M$ (Argoubi and Shirazi-Adl, 1996)

$$k = k_0\left[\frac{e(1+e_0)}{e_0(1+e)}\right]^2 \exp\left[M\left(\frac{1+e}{1+e_0} - 1\right)\right] \tag{4}$$

### 2.3. Model validation

Different validation tests were performed (Table 2):

(i) The RM was loaded at 25 N/s to a maximum of 1000 N in anterior shear. Shear stiffness and load–displacement curve at the transient period were compared with *in vitro* data of Lu et al. (2005). Stiffness was calculated by linear regression analysis of the load-displacement curve from 0 to 250 N according

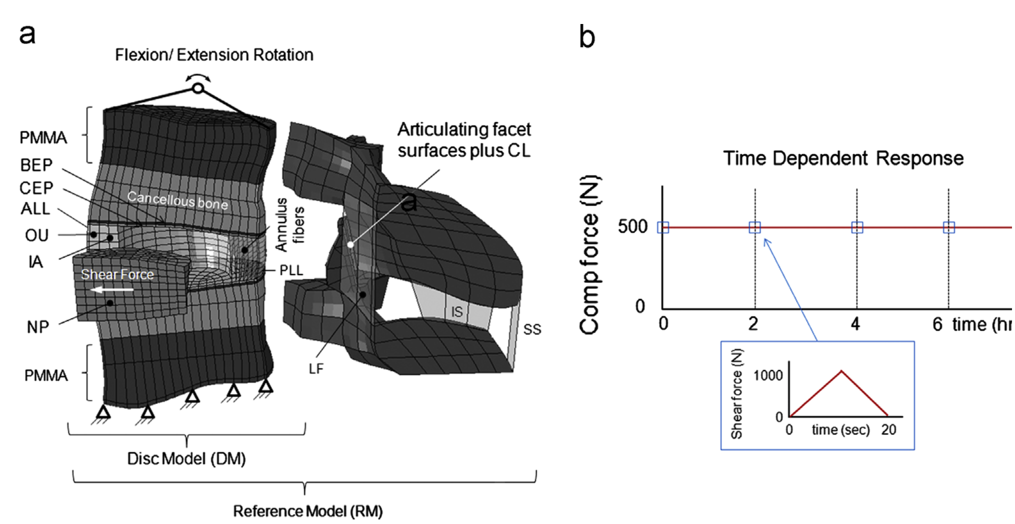


**Fig. 1.** (a) Sagittal symmetric finite element model of the L4-L5 human lumbar spinal segment with nine distinct regions: nucleus pulposus (NP), inner annulus (IA), outer annulus (OA), cartilage endplate (CEP), bony endplate (BEP), cancellous bone, posterior bony elements including the facet articulations, and six ligaments: anterior and posterior longitudinal ligaments (ALL and PLL), ligamentum flavum (LF), supraspinous (SS), interspinous (IS) and capsular (CL) ligaments. According to *in vitro* studies (Skrzypiec et al., 2012), the upper half part of the L4 vertebral body and the lower half part of the L5 vertebral body were embedded in polymethylmethacrylate (PMMA). (b) Time history of the applied diurnal load. An axial compression load of 500 N was applied for a period of 6 h. Directly after the application of compression preloads as well as after 2, 4 and 6 h creep, an anterior shear force was added and removed in 20 s. The shear stiffness was analyzed at the peak of the shear force (after 10 s, see the inlay figure).