View metadata, citation and similar papers at core.ac.uk
brought to you by CORE
provided by IntechOpen

We are IntechOpen, the world's leading publisher of Open Access books Built by scientists, for scientists

5,300
Open access books available

130,000
International authors and editors

155M
Downloads

Our authors are among the

154
Countries delivered to

TOP 1%
most cited scientists

12.2%
Contributors from top 500 universities

CLARIVATE ANALYTICS BOOK CITATION INDEX INDEXED

WEB OF SCIENCE™

Selection of our books indexed in the Book Citation Index in Web of Science™ Core Collection (BKCI)

Interested in publishing with us?
Contact book.department@intechopen.com

Numbers displayed above are based on latest data collected.
For more information visit www.intechopen.com

Finite Element Modeling of the Human Lumbar Spine

Marta Kurutz
Budapest University of Technology and Economics
Hungary

1. Introduction

Finite element (FE) numerical simulation is an effective tool for analyzing phenomena that cannot be clarified by experimental methods, like most of the biomechanical processes, for example the age-related spinal degeneration processes. Moreover, numerical simulation techniques have the potential to reduce costs and to save time during the development of new effective spinal treatment methods or implants. Consequently, there is a need to obtain more and more realistic and correct numerical models for the very complicated structure, the human spine.

In this chapter the FE modeling aspects of the most frequented spinal part, the lumbar spine is presented. After giving a short overview of the anatomy of the lumbar spine, biomechanical effects, loads, internal forces and movements are detailed. Then the three steps of FE modeling procedure, the geometric, material and element/mesh type modeling is discussed, followed by the validation of the complete FE model. Finally, an example for FE numerical simulation closes the chapter.

2. Structural anatomy of the lumbar spine

The lumbar spine is the section of spinal column between the thorax and the sacrum. It consists of five *vertebrae* named L1 to L5 with their posterior elements and articular facet joints, of *intervertebral discs*, *ligaments* and the surrounding *muscles*. In ideal case, the axis of the lumbar spine is a plane curve closed to the section line of the two vertical anatomic planes. The sagittal plane is the vertical plane of symmetry of the body; the frontal or coronal plane is the vertical lateral plane perpendicular to the sagittal one, parallel to the shoulders. The horizontal anatomic plane meets them in the intersection point located in the lumbar spine. The clinical anatomy of the lumbar spine can be studied in the books of Bogduk and Twomey (1987), White and Panjabi (1990), Dvir (2000), Benzel (2001), Adams et al. (2002) and Bogduk (2005).

2.1 Vertebral body, posterior elements, articular facet joints

The lumbar vertebrae are roughly cylindrical with a lateral diameter (width) of 40-50 mm and sagittal diameter (depth) of 30-35 mm. The lumbar vertebrae are thicker anteriorly than

posteriorly resulting in anteriorly convex curvature of the spine known as the lumbar lordosis. Thus, the dorsal body heights are 25-27 mm, the ventral body heights are 26-29 mm.
The *vertebral body* consists of an outer shell of high strength *cortical bone* reinforced internally by the *cancellous bone* as a network of vertical and horizontal narrow bone struts called *trabeculae*. The superior and inferior surface of the vertebral body is covered by the bony endplates of thin cortical bone perforated by many small holes which allow the passage of metabolites from bone to the central regions of the avascular discs.
Towards the upper end of the posterior surface of the vertebral body is a pair of stout pillars of bone called the *pedicles* supporting the *posterior elements*. From each pedicle a skew plate of bone called the *lamina* goes towards the midline where they fuse. The pedicles and the laminae together form the *neural arch*, which together with the posterior surface of the vertebral body encloses a channel, the *vertebral foramen*. The peak-like junction of the two laminae in the midline forms the *spinous process*, while at the lateral junction of each lamina and pedicle starts a long flattened bar of bone called the *transverse process*. At the root of each transverse process, two extensions of bone rise from the lamina: upwards the superior, downwards the inferior *articular process*.
The superior articular processes of a vertebra meet the inferior articular process of the adjacent vertebra above, forming the two symmetrically located synovial joints called *articular facet joints* with an average area of 1.6 cm^2 each. The articular surfaces are approximately vertical in the upper lumbar spine, but are more oblique at the lower part of it.

2.2 Intervertebral discs

The *intervertebral discs* separate the adjacent vertebrae following the geometric measures of the relating vertebrae They are roughly cylindrical with a lateral diameter (width) of 40-45 mm and sagittal diameter (depth) of 35-40 mm. The ratio of disc height to height of the vertebral body is about 1:3 in the lumbar region, so the height of the lumbar disc is about 10 mm. During daily activity, when the discs are under compressive load, water is squeezed out of them, so they lose their height. After bed rest and sleep in the night, water flows back into the discs, so the height of them is restored again.
The structure of intervertebral disc tissue is anisotropic. The disc consists of three components: the gelatinous center, the *nucleus pulposus*, surrounded by the concentrically arranged fibrous layers or lamellae of the *annulus fibrosus*, and the superior and inferior cartilaginous *endplates*.
The nucleus pulposus forms 25-50% of the sagittal cross-area of the disc. It is located more posterior than central in the lumbar spine. The nucleus is a hydrated gel, a semi-fluid mass, an incompressible sphere that exerts pressure in all directions. Although there are significant differences in their structure, there is no clear boundary between the nucleus and the annulus.
The annulus fibrosus consists of 15-25 concentric laminated layers of collagen lamellae tightly connected to each other in a circumferential form around the periphery of the disc. Each lamella consists of ground substance and collagen fibers. Within each lamella the collagen fibers are arranged in parallel, running at an average direction of 30^0 to the discs horizontal plane. In adjacent lamellae they run in opposite directions and are therefore oriented at 120^0 to each other. The outermost fibers called *ligamentous* portion of annulus fibrosus are attached directly to the external rim of vertebral bone, while the internal fibers called *capsular* portion of annulus fibrosus insert into the cartilaginous endplates, forming a continuous

envelope around the nucleus pulposus. Outside the lamellae are mutually connected by the anterior and posterior longitudinal ligaments.

The *endplates* separate the nucleus and annulus from the vertebral bodies. The plates have a mean thickness of 0.6 mm. The cartilaginous endplates cover the superior and inferior surface of the disc, the top and bottom of the nucleus and annulus, binding the disc to its respective vertebral bodies. *Cartilaginous endplates* cover almost the entire surface of the adjacent vertebral bodies, the *bony endplates*, only a narrow rim of bone around the perimeter of the vertebral body is uncovered by cartilage.

2.3 Ligaments

In the lumbar spine seven types of ligaments are distinguished, five of them connect the several parts of posterior elements of vertebrae and two of them connect the vertebral bodies itself.

The *ligamentum flavum* (LF), the most elastic ligament of the spine connects the lower and upper ends of the internal surfaces of the adjacent laminae, closing the gap between the consecutive laminae. The *intertransverse ligaments* (ITL) connect the transverse processes by thin sheets of collagen fibers. The *interspinous ligaments* (ISL) connect the opposing edges of spinous processes by collagen fibers, while the *supraspinous ligaments* (SSL) connect the peaks of adjacent spinous processes by tendinous fibers. The *capsular ligaments* (CL) connect the circumferences of the joining articular facet joints, being perpendicular to the surface of the joints.

The *anterior longitudinal ligament* (ALL) covers the anterior surfaces of the vertebral bodies and discs, attached strongly to the vertebral bone and weakly to the discs. Consequently, it is wider at the bone and narrower at the discs. The *posterior longitudinal ligaments* (PLL) covers the posterior aspects of the vertebral bodies and discs, attached strongly to the discs and weakly to the bone. Consequently, it is wider at the discs and narrower at the bone.

2.4 Muscles

The muscles of the lumbar spine can be distinguished by their location around the spine: *postvertebral* and *prevertebral muscles*. The postvertebral muscles can be divided to *deep*, *intermediate* and *superficial* categories. The prevertebral muscles are the *abdominal muscles*.

The *postvertebral deep muscles* consist of short muscles that connect the adjacent spinous and transverse processes and laminae. The *intermediate muscles* are more diffused, arising from the transverse processes of each vertebra and attaching to the spinous process of the vertebra above. The *superficial postvertebral muscles* collectively are called the *erector spinae*. There are four *abdominal muscles*, three of them encircle the abdominal region, and the fourth is located anteriorly at the midline.

2.5 Functional spinal units or motion segments

A motion segment or functional spinal unit (FSU) is the smallest part of the spine that represents all the main biomechanical features and characteristics of the whole spine. Thus the entire spinal column can be considered as a series of connecting motion segments. The motion segment is a three-dimensional (3D) structure of six degree of statical/kinematical freedom, that is, mathematically, it is 3D in the geometric and 6D in the function space. The FSU consists of the two adjacent vertebrae with its posterior elements and facet joints, and the intervertebral disc between them, moreover the seven surrounding ligaments, without muscles.

3. Biomechanics of the lumbar spine

The spine is a typical mechanical structure. The biomechanics of the spine concerns the mechanical behaviour of the spine as a living load bearing structure subjected to physiologic and other loads. The spinal column is the main load bearing structure of the human musculoskeletal system. It has three fundamental biomechanical functions: (1) to guarantee the load transfer along the spinal column without instability, (2) to allow sufficient physiologic mobility and flexibility, and (3) to protect the delicate spinal cord from damaging forces and motions. The lumbar spine has a distinguished role in these functions: it has the greatest loads, forces and moments, and the greatest mobility at the same time. As a consequence, the lumbar spine is a very complicated compound mechanical structure.

3.1 Loads acting on the lumbar spine

The loads acting on the spine can be divided into two classes: *physiologic* and *traumatic* loads. The physiologic loads due to the common, normal activity of the spine have further classes: *short-term loads* (in flexion, extension), *long-term loads* (in sitting, standing), *repeated or cyclic loads* (in gait, walk), *dynamic loads* (in running, jumping). The traumatic loads generally occurs suddenly, accidentally with great amplitude (in impact, whiplash). The spinal loads based on biomechanical studies are summarized by Dolan and Adams (2001).

The loads can be classified concerning the *origin* of them (gravity, muscles, etc.). Each part of the body is subjected to *gravity loa*d, proportionally to its mass. The compressive gravity load increases downwards, towards the support of the body. Standing upright, the weight of the upper body loads vertically the lower lumbar spine. This load can be multiplied in acceleration, during a fall, or other affect with acceleration or deceleration.

Muscle loads depend on the muscle activity. The muscles are *active tissues*, they can contract, their ability of contraction is governed by the nervous system. The back and abdomen mucsles stabilize the spine in upright standing position, moreover, they prevent the spine from extreme movements. At the same time, since the strongest muscles run parallel and closed to the long axis of the spine, their contraction subjects it to high compressive forces. According to Nachemson (1981) and Sato (1999), during relaxed standing or sitting, the compressive load from the muscles can double the concerning load of body weight. In bending forwards with lifting weights, the back muscles generate very high tensile forces to equilibrate the effects of the vertically acting upper body load and the lifted weight together (Adams et al. (2002)). Due to this extra large tensile force in the back muscles, the lumbar spine is subjected to a high compression.

The *ligaments* are *passive tissues,* they cannot contract, but can sustain high tensile forces being stretched. Stretched passive ligaments store elastic energy that can release and unload the muscles (Dolan et al., 1994).

The *intra-abdominal pressure* decreases generally the spinal compression due to the abdominal muscle activity. Holding the breath increases the intra-abdominal pressure that increases the spinal stability. Wide abdominal belts help to reduce the spinal compressive forces during lifting. It is proved that a belt can reduce sudden unexpected loading events as well.

Ergonomic loads afflict mostly the lumbar spine. By lifting and holding weights the lumbar spine is subjected to high compressive load, depending on the horizontal distance of the load from the lumbar spine. Long-term vibration and cyclical effects may increase the compression in the lumbar spine leading to structural changes and fatigue effects in the tissue of discs and vertebrae.
It has been proved experimentally, that *traumatic overload* of the spine may cause damage in the discs and facet joints. When a disc is loaded beyond its load-bearing capacity, structural damage may occur as a consequence of high-level short-term loads that exceed load tolerance. However, submaximal long-term and repeated loads may also cause disc failure. Although muscles can save the spine from excessive injurious loads and movements, this protection works only if the neural system has time enough to activate the muscles. This time is very short, however, in most cases it is not enough to avoid the injuries. By static loading the protection of muscles generally does work, still injury happens, mainly by coupled torsion and bending in the morning, by sudden lifting, by tired muscles, by vibrational circumstances. By dynamical effects, by car incidents, by whiplash-like loading effects, the arrival of neural information to the muscles is too late. The unexpected dynamic effect of a simple stumble may cause 30-70% larger compression to the lumbar spine. The defensive reaction of the cervical and dorsal muscles can save the life of a person in a car impact.

3.2 Internal forces arising in the lumbar spine

The main internal force acting on the lumbar spine is the *compressive normal force* acting perpendicularly to the middle plane of the discs, causing high compression in the discs. It is accompanied by mainly sagittal and less lateral *shear forces* acting in the middle plane of the discs, causing the slope of the discs to each other. The *moment components* causing the forwards/backwards bending (flexion/extension) and the lateral bending of the spine are the sagittal and lateral *bending moments*, respectively; and the component that causes the spine to rotate about its long axis is the *torque or torsional moment*. The *tensile force* is also a normal force acting perpendicularly to the middle plane of the discs and causing the elongation of it. Although from physiologic loads there is no pure tensile force acting on the spine, since it acts generally to a part of the discs only as a side effect of other internal forces, however, the aim of traction therapies is even to apply pure tensional force to the lumbar spine.
The *compressive force* arisen in the lumbar spine from the body weight depends on the weight of the trunk, head and arms together. Standing upright this vertical weight load is approximately 55-60% of the total body, that is, about 400 N for the standard body weight of 700 N. Taking into account that the lumbosacral disc has approximately 30^0 inclination to the horizontal plane, this force can be decomposed to a 350N compressive normal force and a 200N sagittal shear force. These forces can be doubled by the effect of muscle forces, or by the effect of dynamic loads.
The compressive force in the lumbar spine depends strongly on the posture of the body. In laying posture about 150-250 N, by standing erect about 500-800 N, by sitting erect about 700-1000 N can arise in the lumbar spine (Adams et al. (2002)). During forward bending and lifting weight, the magnitude of the compressive forces increases.
Measuring intradiscal pressure in vivo was made originally by Nachemson (1964, 1981), by sticking a pressure-sensitive needle into the L3-4 disc of volunteers. This ,method has been improved by better technology by Sato (1999), Wilke et al. (1999) and Adams et al. (1996, 2002).

The lumbar *shear force* has never been measured in vivo, thus it can be calculated only, depending on the compressive force and the inclination angle of the actual disc. The shear force is higher in the lower lumbar part due to the higher inclination.
The *bending moments* that play an important role in damaging the discs can be estimated also mainly by mechanical models combined with experimental flexion movement measurements. Adams and Dolan (1991) obtained about 10 Nm bending moment when lifting 100 N weight with bent knees, and 19 Nm with straight knees. Evidently, the lumbar bending moment depends on several variable factors. The back muscles protect the spine from excessive bending during moderate lifting, however, in the case of repeated loading the protective reflex can be eliminated, and the recovery of muscles needs considerable time.
The lumbar spine is subjected to direct *torsion* in some sport or ergonomic activity, accompanied generally by lateral bending. Very little is known about the torsional stresses in the spine in vivo. It is possible to measure the torsional rotation movements in vivo that can be compared with the torque-rotation properties of spine obtained by cadaveric experiments. The upper bound of torque causing damage may be 15-30 Nm in vivo, the safety torque without damage can be 6-12 Nm.
As for the *tensile* force, there are only a few results of measured human spine elongations in pure centric tension, since tensile deformations are analyzed associated with flexion and extension (Bader and Bouten (2000), White and Panjabi (1990)). However, by traction therapies, the lumbar spine is subjected to pure centric tension (Kurutz et al. 2003, Kurutz 2006a, 2006b).

3.3 Mobility of the lumbar spine

The range of spinal movements can be measured both in vivo and in vitro. The spinal motions are described in the space coordinate system, related to the anatomic planes. The spinal movement has six components: three deflections and three rotations. The physiologic movements are the flexion and extension in the sagittal plane, the lateral bending in the frontal plane and the rotation around the long axis of the spine. The spinal motions are generally characterized by three parameters: (1) the *neutral zone* in which the spine shows no resistance, (2) the *elastic zone* in which the spinal resistance works, and (3) the *range of motion,* the sum of the two latter zones. The reference position is the erect standing. The range of motion of a general lumbar segment is about $12-16^0$ for flexion/extension, increasing from L1-2 to L4-5; about 6^0 for lateral bending, and about 2^0 for axial torsion. The neutral zone is about $2-3^0$ for all physiologic motions. The displacement of a motion segment in the neutral zone is about 0.3 mm for tension and compression, and about 0.8 mm for shear. The range of motion is about 0.1-1.9 mm for tension, compression and shear.
The mobility of the spine depends on several factors. It depends first of all on the state of the intervertebral discs: the geometry, the stiffness, the fluid content, the degeneration and aging of it. The smaller its diameter and the larger its height is, the larger its mobility, its any range of motion is. Because of the greater proportional height of the discs, the lumbar region of the spine has greater mobility than the thoracic spine. The range of motion is influenced also by the state of ligaments, the articular facet joint and the posterior bony elements. Viscoelastic properties of discs and ligaments also have an effect on the mobility.

3.4 Degeneration of the lumbar spine

Degeneration means a specific injurious change in the composition, structure and function of the spine. Degeneration is a general terminology, it may be divided basically into two main classes: (1) the long-term *age-related degenerations* and (2) the shorter-term *environmental degenerations* caused by mechanical, chemical, electromagnetic or any other environmental effects. Among the latter category, there is a large class of sudden short-term degenerations caused by unexpected mechanical effects, like a sudden overload or wrong movements, car accidents or other traumatic loads, leading to mechanical damage of the spine.

Age-related degenerations may lead to annular tears or disc prolapse and herniation or osteoporotic failures in the trabecular bone. The forms of environmental degeneration depend on the given special causing effects. Compression overload may cause vertebral failures, endplate fractures cancellous bone collapse, and internal disc disruption, annulus buckling, due to some alarming events, heavy weight lifting in a rapid manner, accidents involving falls, collisions. Shear overload may lead to articular joint damage and fracture, spondylolysis, caused by a heavy back pack or an excessively lordotic posture. Torsion overload can cause impact of articular joint again and anterior annulus tears, caused by a fall or some sports. Forward bending may lead to intervertebral ligament sprains, caused by putting on a shoe, or bending forward suddenly after a long period of flexed posture like car driving, during repetitive bending and lifting, Backwards bending may yield impact of neural arch and posterior disc bulging, caused by an overhead manual work or some sports. Lateral bending may cause ligament sprains and articular joint impaction, caused by special movements during daily activity. However, highest risk of mechanical damage and the most dangerous injuries occur during the combination of the listed mechanical effects that are generally in multiple combinations with each other. Bending and compression combined with torsion has the highest risk to spinal damage. Unexpected, sudden load effects, vibration, repetitive load and fatigue may increase the risk for damage (Adams et al., 2002).

Any age-related or environmental change in the structure of discs or vertebrae leads to the change in the load-bearing capacity of the lumbar segment. Osteoporosis of vertebrae, or decreasing water-binding capacity of nucleus, or calcification of the fibro-cartilaginous endplates, or a disc prolapse of sudden loading or a wrong movement, equally leads to metabolic disorders and biomechanical defects, to limited ability of the required mechanical function and loading tolerance of segments (Adams et al., 2002).

4. Biomechanics of the elements of the lumbar functional spinal unit

The three-dimensional FSU has six force and six motion components that occur in combination with each other. These general force and motion systems depend highly on the mechanical properties, stiffness or flexibility, or load bearing capacity of each structural component of the motion segment.

4.1 Biomechanics of the vertebral body and the articular facet joints

Lumbar vertebral bodies resist most of the compressive force acting down along the long axis of the spine. Most of this load must resisted by the dense network of trabeculae, and less by the cortical shell. Namely, in the load transfer of axial compression, the nucleus of the disc pressurizes the cartilaginous endplates to bulge inward the cancellous core of vertebrae, when the trabecular bone columns start to buckle due to the excessive load, consequently, first the trabecular bone fails during compression. During this process, radial stresses occur in the endplates, causing cracks in it, to allow the nucleus to bulge also into the vertebral body. Thus, the state of the cancellous bone is the main factor of failure tolerance of vertebrae (McGill, 2000). Moreover, the cancellous bone of vertebrae acts as shock absorber of the spine in accidental injurious effects.

The load bearing capacity of vertebrae depends mainly on the geometry, mass, bone mineral density (BMD) and the bone architecture of the vertebral cancellous bone, which are in correlation with aging, sex and degeneration. *Mosekilde* (2000) demonstrated that age is the major determinant of vertebral bone strength, mass, and micro-architecture. Some papers consider the effect of aging and sex on the compressive strength characteristics of vertebrae. There is a significant decrease of vertical compressive strength and load baring capacity of vertebral trabecular bone with the development of osteoporosis during aging, occurring in different life periods for men and women (Duan et al. 2001, Keaveny and Yeh 2002). Analysis of regional inhomogeneity inside the vertebral body showed that the weakest part is the central region of the vertebral body *(*Gong et al. 2005, Briggs et al. 2004. Banse et al. 2001).

The posterior elements of vertebrae (pedicles, laminae, spinous and transverse processes and facet joints) have also important role in the load bearing capacity and mobility of segments. Facet joints work as typical contact structures governed by unilateral conditions, limiting the spinal movements, extension, lateral bending and axial torsion. Failure of the posterior elements, together with facet damage, leading to spondylolisthesis, is generally caused by anteroposterior shear forces. The articular facet joints stabilize the lumbar spine in compression, and prevent excessive bending and translation between adjacent vertebrae. In this manner they are able to protect the disc. Lumbar facet joints are able to resist forces acting perpendicular to the vertical articular surfaces, approximately in the plane of the disc. Thus, they limit the range of axial rotation, with great contact stresses occurring in the joint surfaces.

4.2 Biomechanics of the intervertebral disc

The intervertebral discs provide the compressive force transfer between the two adjacent vertebrae, at the same time, they allow the intervertebral mobility and flexibility. The arrangement of the collagen fibers in the annulus fibrosus is optimal for absorbing the stresses generated by the hydrostatic compression state of the nucleus pulposus in axial loading of the disc, moreover, they play an important role in restricting axial rotation of the spine.

Axial compressive stiffness is higher in the outer and posterior regions than in the inner and anterior regions. Tensile stiffness is higher in the anterior and posterior part than in the lateral and inner regions. Thus, the inner annulus near the nucleus seems to be the weakest area of annulus, and the outer posterior part the strongest region.

In sustained loading the spine shows viscoelastic features. In quasi-static compression the disc creep is 5-7 times higher than the creep in the bony structures of the segment. Thus, the main factor of segment viscosity is the disc, mainly the disc annulus. The creep of the disc depends on the fluid content of it. A 3 hours long 1200 N compressive loading yields a 10% decrease in disc height and 5-13% increase in the sagittal diameter (Adams and Hutton, 1983, McNally and Adams, 1992) mainly due to fluid loss, similarly to the diurnal variation, namely the effect of overnight bed rest with fluid recovery.

The load bearing capacity of segments is mainly influenced by the degeneration state of the disc. Degeneration means an injurious change in the function and structure of the disc, caused by aging or by environmental effects, like mechanical overloading (Adams et al., 2000). Degeneration of FSU starts generally in the intervertebral discs. Changes to any tissue property of the disc markedly alter the mechanics of load transfer and stability of the whole segment (Ferguson and Steffen, 2003). The first age-related changes of disc occur within the nucleus.

Long-term age-related degeneration of the disc is manifested in the loss of hydration, a drying and stiffening procedure in the texture of mainly the nucleus (McNally and Adams 1992; Adams et al. 2002; Cassinelli and Kang, 2000). The functional consequences of aging are that the nucleus becomes dry, fibrous and stiff. The volume of nucleus and the region of hydrostatic pressure of it decrease, consequently, the compressive load-bearing of the disc passes to the annulus. However, the annulus becomes weaker with aging, so the overloading of it can lead to the inward buckling of the internal annulus, or to circumferential or radial tears, fiber break in the annulus, disc prolapse or herniation, or to large radial bulging of the external annulus, reduction of the disc height, or moreover, to endplate damages (Natarajan et al., 2004). The main cause of all these problems is that while the healthy disc has a hydrostatic nucleus, it becomes fibrous during aging, being no longer as a pressurized fluid.

Short term sudden degeneration or damage may yield the sudden loss of hydrostatic compression in nucleus, accompanied or due to some other failures mentioned above. Several recent studies concluded that light degeneration of young discs leaded to instability of lumbar spine, while the stability restored with further aging (Adams et al., 2002).

4.3 Biomechanics of the ligaments and muscles

The ligaments are passive tissues working only against tension. The primary action of the spinal ligaments lying posterior to the centre of sagittal plane rotation is to protect the spine, by preventing excessive lumbar flexion. However, during this protection the ligaments may compress the discs by 100% or more. Indeed, the effectiveness of a ligament, its contribution to the integrity of the spine depends mainly on the moment arm through which it acts.

The most elastic ligament, the ligamentum flavum being under pretension throughout all levels of flexion prevents any forms of buckling of spine. The interspinous and supraspinous ligaments may protect against excessive flexion. The capsular ligaments of facet joints restrict joint flexion and distraction of the facet surfaces of axial torsion.

The failure strength of the lumbar ligaments are about 450 N for the anterior longitudinal, 330 N for the posterior longitudinal, 220 N for the capsular, 120 N for the interspinous ligaments and 280 N for the ligamentum flavum (Benzel, 2001).

The muscles and the neuromuscular controls are required (1) to provide dynamic stability of the spine in the given activity and posture, and (2) to provide mobility during physiologic activity, moreover (3) to protect the spine during trauma in the post-injury phase. Two mechanical characteristics are necessary to provide these physiologic functions: (1) the muscles must generate forces isometrically and by length change, and (2) they must increase of the stiffness of the spinal system.

4.4 Biomechanics of the functional spinal unit

The mechanical behaviour of the FSU depends on the physical properties of its components, mainly on behaviour of the intervertebral disc, ligaments and articular facet joints.

The average *load tolerance* of lumbar segments under quasi-static loading is about 5000 N for compression, 2800 N for tension, 150 N for shear and 20 Nm for axial rotation (Bader and Bouten, 2000).

Flexibility of the FSU is the ability of the structure to deform under the applied load. Inversely, the *stiffness* is the ability to resist by force to a deformation. The stiffness of the spinal segments increases from the cervical to the lumbar regions for all loading cases. In lumbar region the stiffness is about 2000-2500 N/mm for compression, 800-1000N/mm for tension, 200-400 N/mm for lateral and 120-200 N/mm for anterior/posterior shear. The rotational stiffness is about 1.4-2.2 Nm/degree for flexion, 2.0-2.8 Nm/degree for extension, 1.8-2.0 Nm/degree for lateral bending and 5 Nm/degree for axial torsion (White and Panjabi, 1990, Bader and Bouten, 2000). The stiffness of the lumbar spine depends on the age and degeneration. In advanced degeneration the stiffness is higher. The stiffness is influenced by the viscous properties of the segments and the load history as well.

5. Geometrical modeling of lumbar functional spinal unit

Geometrical modeling of the FSU needs precise geometrical data of the real object; it must follow the anatomy of the segment. Beside the topology, additional data such as volume density, surface texture, etc. are needed. Different methods of acquisition of geometrical data can be used, like scanners, computer tomography, or magnetic resonance imaging methods.

In FE modeling the vertebral body, its cortical shell, cancellous core, posterior bony elements and the bony endplates are generally distinguished. For the thickness of the vertebral cortical wall Lu et al. (1996) used 1.5 mm, and for the thickness of the cartilaginous endplates considered 1 mm. Baroud et al. (2003) applied 1 mm cortex and 0.5 mm endplate thickness. For the thickness of the cartilage layer of facet joint Schmidt et al., (2009) considered 0.2 mm.

In FE modeling the intervertebral disc, its nucleus, annulus ground substance, annulus fibers and the cartilaginous endplates are generally distinguished. For the volumetric relation between annulus and nucleus, ratio 3:7 is generally used for the lumbar part L1-S1 (Goto et al., 2002, Moramarco et al., 2010). Chen et al. (2001) considered 30-50% of the total disc area in cross section as the nucleus and the rest of the region as the annulus. Lu et al. (1996) applied 38% nucleus area ratio to the total disc cross-sectional area, based on the measurements of a normal disc. Baroud et al. (2003) assumed the nucleus to occupy the 43% of the total disc volume. The diameter length of the disc from anterior to posterior end is about 36 mm, the lateral length is about 44 mm. For the orientation of annulus fibers Rohlmann et al. (2007) considered to the mid cross-sectional area of the disc under alternating direction of about 30^0 and 150^0.

6. Material modeling of lumbar functional spinal unit

Since FSU is a highly heterogeneous compound structure, the material modeling must be related to the components of it. First the material models of the healthy components are considered.

6.1 Material models of the vertebral body and the articular facet joints

The high strength *vertebral cortical shell* is generally considered linear elastic isotropic or transversely isotropic, orthotropic material, seen in Table 1. Vertebral *cancellous bone* is modeled generally by linear elastic isotropic or transversely isotropic or orthotropic material, seen in Table 2.

Vertebral cortical bone				
Material model	**E** [MPa]	ν	**G** [MPa]	**References**
linear elastic, isotropic	5 000	0.3		Rohlmann et al. 2006b, Zander et al. 2006,
linear elastic, isotropic	10 000	0.3		Argoubi, Shirazi-Adl 1996, Kumaresan et al. 1999, Rohlmann et al. 2006a, 2006c, 2007,
linear elastic, isotropic	11 300	0.2		Little et al. 2008,
linear elastic, isotropic	12 000	0.3		Shirazi-Adl et al. 1984, Cassidy et al. 1989, Lavaste et al. 1992, Goel et al. 1995a, 2002, Lu et al. 1996b, Smit et al. 1997, Wang et al. 2000, Chen et al. 2001, 2008, Baroud et al. 2003, Zhong et al. 2006, Denoziere and Ku, 2006, Williams et al. 2007, Ruberté et al. 2009, Zhang et al. 2009, Kurutz and Oroszváry 2010
linear elastic transversely isotropic	11 300 11 300 22 000	0.48 0.20 0.20	3 800 5 400 5 400	Lu et al. 1996a, Schmidt et al. 2009,
linear elastic, transversely isotropic	8 000 8 000 12 000	0.40 0.23 0.35	2 857 3 200 3 200	Noailly et al. 2005, 2007, Malandrino et al. 2009,
poroelastic	10 000	0.3		Ferguson et al. 2003,

Table 1. Material properties for the FE models of lumbar vertebral cortical bone

Vertebral cancellous bone				
Material model	**E** [MPa]	ν	**G** [MPa]	**References**
linear elastic, isotropic	10	0.2		Shirazi-Adl et al. 1984, Cassidy et al. 1989, Lu et al. 1996b, Smit et al. 1997, Goel et al. 1995a, 2002, Wang et al. 2000, Chen et al. 2001, 2008, Denoziere and Ku 2006, Zhong et al. 2006, Ruberté et al. 2009,
linear elastic, isotropic	50 81 140 100 100 150 500	0.2 0.2 0.2 0.29 0.3 0.3 0.2		Rohlmann et al. 2006a, Baroud et al., 2003, Little et al., 2008, Zhang et al. 2009. Lavaste et al., 1992, Kurutz and Oroszváry, 2010, Rohlmann et al. 2006b, Zander et al., 2006,
poroelastic	100	0.2		Argoubi and Shirazi-Adl 1996, Williams et al. 2007
linear elastic transversely isotropic	200 140	0.45 0.315		Rohlmann et al. 2006c, 2007,
linear elastic transversely	140 140	0.45 0.176	48 77	Noailly et al. 2005, 2007, Malandrino et al. 2009,

isotropic	250	0.315	77	
linear elastic transversely isotropic	140 140 200	0.45 0.32 0.32	48.3 48.3 48.3	Lu et al. 1996a, Schmidt et al. 2009,

Table 2. Material properties for the FE models of lumbar vertebral cancellous bone

The high strength *bony endplate* of vertebrae and the lower strenght *cartiliginous endplate* of disc can hardly be distinguished in FSU. The authors generally give information about it when specifying material properties. Table 3 shows the generally applied material moduli of the endplates.

Endplate			
Material model	**E [MPa]**	ν	**References**
bony, linear elastic, isotropic	12 000 1000 1000 500 500 100	0.3 0.3 0,4 0.4 0.3 0.4	Baroud et al. 2003, Noailly et al. 2005, 2007, Chen et al. 2001, 2008, Lavaste et al. 1992, Zhang et al. 2009, Kurutz and Oroszváry 2010,
cartiliginous, linear elastic, isotropic	24	0.4	Shirazi-Adl et al. 1986, Goel et al. 1995a, 2002, Noailly et al. 2005, 2007, Wang et al. 2000, Zhong et al. 2006, Ruberté et al. 2009, Lu, et al. 1996, Schmidt et al, 2009
*cartiliginou*s, poroelastic	5 5 20	0.1 0.17 0.4	Argoubi and Shirazi-Adl 1996, Malandrino et al. 2009, Williams et al. 2007,
bony, outer: intermediate: central:	12 000 6 000 2 000	0.3 0.3 0.3	Denoziere and Ku 2006

Table 3. Material properties for the FE models of lumbar vertebral and disc endplates

The *posterior bony elements* are considered linear elastic isotropic material, generally by the same Young's modulus E=2500 MPa and Poisson's coefficient ν =0.25 or ν =0.2.
The *articular facet joints* are considered as unilateral frictionless connections transmitting only compressive forces with an initial gap of generally 0.5 mm (Rohlmann et al. 2006a, Chen et al., 2008, Zhang et al., 2009); or 0.6 mm of a nonlinear frictionless contact problem Schmidt et al., 2009); or assuming soft contact with exponentially increasing contact force with decreasing contact gap (Sharma et al., 1995, Rohlmann et al. 2007). Zhong et al. (2006) used 1 mm initial gap for the surface-to surface frictional contact with friction coefficient 0.1. Little et al. (2008) used a finite sliding frictionless tangential relationship with softened contact in the normal direction that means exponentially increasing contact stresses with initial gap of 0.1 mm.

6.2 Material models of the intervertebral disc

The *intervertebral disc* is the most critical component of the spine in both its mobility and load bearing ability, therefore its FE modeling has a great importance.

6.2.1 Material models of the nucleus pulposus

Nucleus pulposus is the most important element in the compressive stiffness of the disc: the hydrostatic compression in it guarantees the stability of the whole disc and segment. The healthy young nucleus is generally modeled as an incompressible fluid-like material, seen in Table 4.

Nucleus pulposus			
Material model	**E** [MPa]	ν	**References**
Fluid-like solid, linear elastic, isotropic	1	0.499	Shirazi-Adl et al. 1984, 1986, Goel et al. 1995a, Chen et al. 2001, Zhong et al. 2006, Denoziere and Ku 2006, Zhang et al. 2009, Ruberté et al. 2009, Kurutz and Oroszváry 2010,
	4	0.499	Shirazi-Adl et al. 1984, Lavaste et al. 1992, Goel et al. 1995b, Fagan et al, 2002.
	10	0.4	Chen et al., 2008,
Incompressible fluid			Lu et al. 1996, Little et al. 2008, Zander et al. 2006, Rohlmann et al. 2006a, 2006b, 2006c
Quasi incompressible			Rohlmann et al. 2007
Hyperelastic, neo-Hookean			Moramarco et al. 2010,
Mooney-Rivlin incompressible			Smit et al. 1997, Noailly et al. 2007, Baroud et al. 2003, Schmidt et al. 2007, 2009,
Poroelastic	varied	0.17	Malandrino et al. 2009,
	1	0.45	Williams et al. 2007,
	1.5	0.1	Argoubi and Shirazi-Adl 1996,
	1,5	0.17	Ferguson et al. 2004,
Viscoelastic solid	2	0.49	Wang et al. 2000,
Osmoviscoelastic	0.15	0.17	Schroeder et al. 2006

Table 4. Material properties for the FE models of lumbar disc nucleus

6.2.2 Material models of the annulus fibrosus

Annulus fibrosus represents a typical composite-like material with a ground substance and fiber reinforcements of many layers, seen in Table 5.

Annulus fibrosus					
	Ground substance		**Fibers**		
Material model	**E** [MPa]	ν	**E** [MPa]	ν	**References**
Linear elastic, isotropic matrix, tension only elastic fibers	4	0.4	500	-	Lu et al. 1996,
	4	0.45	500	0.3	Fagan et al. 2002)
	4	0.45	400/500/300	0.3	Kurutz and Oroszváry 2010,
	4.2	0.45	450	-	Zhong et al. 2006,
	4.2	0.45	450	0.3	Shirazi-Adl et al. 1984, 1986,
	4.2	0.45	175	-	Chen et al. 2001,
	4.2	0.45	360/420/485/550	0.3-	Denoziere and Ku, 2006,
	2	0.45	500	-	Lavaste et al. 1992,

	2 8 10 4.2 4.2	0.45 0.45 0.4 0.45 0.45	500 360/420/485/500 360/385/420/440/495/550 nonlinear 500	 - 0.3 	Goel et al. 1995b, Baroud et al. 2003, Chen et al. 2008, Goto et al. 2002, Zhang et al. 2009,
Hyperelastic matrix, nonlinear outwards stiffening fibers	3.15	0.45			Schmidt et al. 2007, 2009, Rohlmann et al. 2006a, 2006b, 2006c, 2007, Ruberté et al. 2009, Noailly et al. 2007, Moramarco et al., 2010,
Hyperelastic matrix, linear elastic tension only fibers			500	0.3	Little et al. 2008,
Viscoelastic matrix, nonlinear elastic fibers	8	0.45			Wang et al. 2000,
Poroelastic matrix, nonlinear elastic fibers	2.5 2.5	0.4 0.1			Williams et al. 2007, Argoubi and Shirazi-Adl 1996,
Poroelastic matrix, linear elastic fibers	2.5	0.17	60	0.33	Ferguson et al. 2004,

Table 5. Material properties for the FE models of lumbar disc annulus

6.3 Material models of the ligaments

Numerical modeling of *ligaments*, as typical exponentially stiffening soft tissues is not a simple task. Generally, the seven ligaments are incorporated to the FE models as tension only elements. In contrast to its strong nonlinear behaviour (White and Panjabi, 1990), most of the reported FEM studies have adopted linear elastic models (Lavaste et al., 1992, a et al., 1995, Zhong et al, 2006, Chen et al, 2008), but bilinear models (Pintar et al., 1992, Goel et al, 1995a, Chen et al., 2001, Goto et al., 2002, Denoziere and Ku, 2006, Moramarco et al, 2010), moreover, trilinear approaches are also used (Pintar et al, 1992, Ruberté et al, 2009), seen in table 6.

Lumbar Ligaments								
Ligaments	E_1 [MPa]	ε_1 [%]	E_2 [MPa]	ε_2 [%]	E_3 [MPa]	ε_3 [%]	**CS area** [mm²]	**References**
ALL PLL LF ITL CL ISL SSL	20 20 19.5 58.7 32.9 11.6 15						63.7 20 40 3.6 60 40 30	Goel et al. 1995a, Zhong et al. 2006
ALL PLL	 70						 20	

LF	50						60	Chen et al. 2008,
ITL	50						10	
CL	20						40	
ISL	28						35.5	
SSL	28						35.5	
ALL	7.8	12	20				63.7	Goel et al. 1993,
PLL	10	11	50				20	1995a, Denoziere
LF	15	6.2	19				40	and Ku 2006,
ITL	10	18	59				1.8	
CL	7.5	25	33				30	
ISL	8	20	15				30	
SSL	10	14	12				40	
ALL	7.8	12	20				63.7	
PLL	10	11	20				20	
LF	15	6.2	19.5				40	
ITL	10	18	58.7				1.8	Chen et al. 2001,
CL	7.5	25	32.9				30	
ISL	10	14	11.6				30	
SSL	8	20	15				40	
ALL	7.8	12	20				32.4	
PLL	1	11	2				5.2	Pintar et al. 1992,
LF	1.5	6.2	1.9				84.2	Goel et al, 1995a,
ITL	10	18	59				1.8	Goto et al. 2002,
CL							43.8	Moramarco et al,
ISL							35.1	2010,
SSL	3	20	5				25.2	
ALL	12.6	8	15.6				32.5	
PLL	27.1	7	40	25	31.6	38	5	
LF	24	8	40	20	36	25	91.6	Pintar et al. 1992,
ITL	125	8	313				2	Ruberté et al,
CL	7.5	25	12.7				51.2	2009,
ISL	4.15	20	11.4				34	
SSL	4.15	20	11.4				34	

Table 6. Material properties for lumbar ligament FE models

Here the transition strains $\varepsilon_1, \varepsilon_2$ and ε_3 separate the concerning Young's moduli of the polygonal stress-strain function. The generally used – very divergently applied - cross sectional areas of each ligament are also illustrated in Table 6. Several further nonlinear FEM models of ligaments can be found in the literature (Shirazi-Adl, 1986a, 1986b, Wang et al., 2000, Zander et al., 2006, Noailly et al., 2007, Rohlmann et al., 2006a, 2006b, 2006c, 2007, Williams et al., 2007, Schmidt et al., 2009). Based on several FE studies Eberlein et al., (2004) summarize the ligament models applied in FEM analyses and suggests a membrane model with a new constitutive equation as the special case of the equation obtained for the annulus fibrosus.

7. Material and geometrical modeling of the degenerated lumbar functional spinal unit

Aging type degeneration starts generally in the nucleus. A healthy young fluid-like nucleus is in a hydrostatic compression state. During aging, the nucleus loses its incompressibility and becomes even stiffer and stiffer, changing from fluid to solid material. This kind of nucleus degeneration can be modeled by decreasing Poisson's ratio with increasing Young's modulus (Kurowski and Kubo, 1986; Kim et al., 1991). This behavior is generally accompanied by the stiffening process of the disc as a whole and by the volume reduction of the nucleus and volume extension of the annulus, furthermore, height reduction of the disc. Moreover, at the same time, annulus tears or internal annulus buckling, or break of the annular fibers, damage and crack or rupture of endplates, osteoporotic defects of vertebral cancellous bone can happen. Consequently, modeling age-related degeneration of FSU is a compound task; it must be done in its progress, relating to a lifelong process.

In contrast to the age-related degeneration, the nucleus may lose its incompressibility without any stiffening and volume change process. due to a *sudden unexpected traumatic load* effect. In this case the nucleus may quasi burst out and the hydrostatic compression may suddenly stop in it. This kind of nucleus degeneration can be modeled by suddenly decreasing Poisson's ratio with unchanged Young's modulus of nucleus (Kurutz and Oroszváry, 2010). This behaviour is generally caused or accompanied by the tear or buckling of the internal annulus, break of the annular fibers, fracture of endplates, or collapse of vertebral cancellous bone, depending on the age in which the accidental event happens. Namely, accidental failures can happen in a young disc, as well, or in any age and aging degeneration phases. These effects can be modeled by sudden damage of tissues of the concerning components of the segment. In contrast to the long term aging degeneration, these kinds of damage instability occurs suddenly, generally due to a mechanical overloading (Acaraglou et al., 1995).

Rohlmann et al. (2006c) have developed a FE model of a lumbar motion segment of different grades of age-related disc degenerations to simulate the effect of degeneration on the biomechanical behaviour of the segment. They introduced three grades of disc degeneration: mild, moderate and severe degenerations. Compared to the healthy disc, the three grades have 20, 40 and 60% less disc height, respectively. Parallel to the disc height reduction, the length of the annulus fibers was also reduced, compensated by offsetting their nonlinear stiffness curves. The facet orientation was also changed with disc height reduction. The compressibility of nucleus was increased with the loss of fluid-like behaviour from 0.0005 to 0.15 mm^2/N, by using linear interpolation for the different grades of degeneration (mild: 0.0503, moderate: 0.0995 mm^2/N). It was assumed that the disc degeneration has no effect on the material properties of the annulus fibrosus. By analyzing a healthy and a slightly degenerated lumbar spine Rohlmann et al. (2007) assumed compressible nucleus with the compressibility of 0.0005 mm^2/N for healthy, and 0.0503 mm^2/N for mildly degenerated case.

Schmidt et al. (2007) verified the hypothesis that with increasing disc degeneration, the internal pressure and strains of the disc decrease, therefore, the risk of disc prolapse decreases. They assumed mildly, moderately and severely degenerated disc with 16.5, 49.5 and 82.5% reduced height, respectively, and by supposing increasing osteophytes formations with progressing degeneration (1.5, 4.5 and 7.5 mm, respectively). Parallel to the disc height reduction, the facet orientation was also changed, compared to the parallel position of the surfaces (0^0), the angle between them increased (0.8, 1.9 and 2.9^0). Moreover, it was supposed that the endplate flattens with progressing disc degeneration. By assuming 0% for the healthy endplate curvature and 100% for the planar endplate, the degeneration percents followed the percents of the disc height reduction. Since the disc bulging increase with the progress of degeneration, and however, there were no data available in the literature, it was assumed the same ratio for bulging increase as for disc height decrease. In the length of the annulus fibers, the results of Rohlmann et al (2006c) were used. The Young's modulus of the nucleus was increased during degeneration from the value of the healthy nucleus to the value of the annulus ground substance by supposing that the nucleus and the annulus become structurally similar with increasing disc degeneration, and that the disc degeneration has no effect on the material properties of the annulus fibrosus.

Ruberté et al. (2009) simulated aging degeneration of L4-5 lumbar segment by introducing three grades of degeneration: healthy, mild and moderate phases; by decreasing disc height (12.0, 10.2 and 8.0 mm) and nucleus area (388, 269 and 101 mm^2) with increasing annulus area (731, 850 and 1022 and mm^2), and by modifying the material properties of annulus ground substance (Mooney-Rivlin C_1/C_2: 0.2/0.05, 0.4/0.1 and 0.9/0.23) and nucleus (Young's modulus/Poisson's coefficient: 1.0/0.49, 1.26/0.45 and 1.66/0.4). The nucleus area was reduced by following the stress profilometry results of Adams et al. (1996). Material properties were taken from the literature and from the results of Umehara et al. (1996), Iatridis et al. (1997) and Elliott and Setton (2001). Umehara et al. (1996) measured experimentally the distribution of the compressive elastic moduli in the lumbar intervertebral disc in term of degeneration. The distribution of elastic moduli in normal discs was symmetric about the midsagittal plane, degenerated discs showed irregular distributions of elastic moduli. The elastic moduli of the degenerated nucleus were higher than those in normal discs.

Zhang et al. (2009) applied healthy, and two degenerated grades in modeling L4-5 motion segment. For grade 1 the elastic modulus of the disc nucleus was two times the elastic modulus of the annulus in the intact model and the Poisson's ratio was adopted to be the same as that of the annulus, and the disc height was reduced by 20%, following Kim et al. (1991), Iatridis et al. (1997) and Kumaresan et al. (2001). For grade 2, in addition, the elastic modulus of the annulus was doubled, and the annulus fiber volume was reduced by 25%, and the disc height was reduced by 40%, by considering Rohlmann et al. (2006c).

Kurutz and Oroszváry (2010) introduced five grades of age-related degeneration from healthy (1) to fully degenerated (5) cases, modeled the loss of hydrostatic state in nucleus by decreasing Poisson's ratio (nu=0.499, 0.45, 0.40, 0.35, 0.30, respectively), accompanied by nucleus stiffening modeled by increasing Young's modulus (E=1, 3, 9, 27, 81 MPa, respectively). Simultaneously, in the annulus matrix a gradual increase (E=4.0, 4.5, 5.0, 5.5, 6.0 MPa), while in the vertebral cancellous bone (E=150, 125, 100, 75, 50 MPa) and endplates (E=10, 80, 60, 40, 20 MPa) a gradual decrease of Young's modulus were considered with aging. The age-related change of tensile moduli of annulus ground substance and nucleus equally (E=0.4, 1.0,1.6, 2.2, 2.8 MPa) were modeled by using a parameter identification method based on the in vivo measured lumbar disc elongations by Kurutz et al. (2003, 2006a, 2006b).

Most of the geometric models of the degenerated disc apply the reduction of the disc height. Lu et al. (1996) by using FE simulation concluded that variation in disc height had a significant effect on the axial displacement, the posterolateral disc bulge and the tensile stress in the peripheral annulus fibers, but the influence on the intradiscal pressure and the longitudinal stress distribution at the endplate-vertebra interface was minimal.

Permeability is a key factor in poroelastic FE analysis, representing the ability of interstitial fluid flow within the tissues. Experimental studies have shown a large variation of this parameter. Gu et al. (1999) and Johannessen and Elliott (2005) correlated anisotropic nucleus and annulus permeabilities with disc degeneration.

Gu et al.(1999) concluded that the fluid transport within a disc is crucial to its viscoelastic behavior, fluid pressure redistribution, and cell nutrition, in viscoelastic behaviors of healthy and degenerate discs as well as the biomechanical etiology of disc failure. The hydraulic permeability of human non-degenerated annulus fibrosus is direction-dependent, namely, anisotropic, with the greatest permeability in the radial direction. With disc degeneration, the radial permeability of annulus decreases, mainly because of the decreased water content, and the axial and circumferential permeability coefficients increase, mainly because of the structural change, leading to more isotropic permeability behavior for more degenerated discs.

Johannessen and Elliott (2005) by measuring the biphasic compressive material properties of normal and degenerate human nucleus pulposus tissue in confined compression concluded that swelling is the primary load-bearing mechanism in both nondegenerate and degenerate nucleus pulposus. Degeneration produced significant decreases in swelling stress, while permeability increased with degeneration.

Malandrino et al. (2009) aimed to study the poromechanical responses of the L3-4 disc FE model under compression, flexion and rotation, by varying the Young's moduli and permeabilities of the tissues of the main components of disc. This may represent a typical degeneration process. They considered the Young's modulus for the healthy nucleus as 1 MPa, for the healthy annulus 2.56 MPa; and for the fully degenerated case for the nucleus 1.66 MPa and for the annulus 12.29 MPa (Iatridis et al., 1998, Natarajan et al., 2006). Four hydraulic permeabilities were chosen, for annulus, nucleus, cartilaginous endplate and for the trabecular bone. This allowed studying the significance of fluid-solid interaction in the disc.

8. Loads in finite element simulation

Loads on lumbar spinal motion segments in FE modeling depend on the aims of the analysis. The segment is generally supported rigidly along the inferior endplate of the lower vertebra, thus, the loads are generally applied on the superior endplate of the upper vertebra.

The loads can be applied as *static or dynamic loads*. Constant static loads or *incrementally changing quasi-static loads* are generally applied in lumbar spine analyses. The basic loading types are the *force or displacement type loads*, in a *load or displacement controlled device*, in a *load history analysis*.

Chen et al. (2001) in analyzing the adjacent segment syndrome of a rigid fixation used equally 10 Nm for flexion, extension, lateral bending and axial torsion under 150 N preload.

Goto et al (2002) investigated the numerical analysis of lumbar vertebrae of L4-5 segment by using intradiscal pressure in nucleus to establish the model, and by applying incremental loading device. Compressive loading was performed to 294 N in ten steps, then flexion and extension loads of 15 Nm was applied in 15 steps. The intradiscal pressure was set as 1.32 MPa for flexed and for standing position; 0.6 MPa for extended position, and zero pressure was assumed for degenerated disc model.

Baroud et al. (2003) applied displacement control in load history FE analysis of vertebroplasty of the L4-5 segment, by applying quasi-static compression load of 2.8 mm in steps of 0.2

mm. Ferguson et al. (2004) analyzed the fluid flow within the disc by simulating the diurnal loading cycle consisting of an 8 hours resting period, followed by a 16 hours constant compressive load equivalent to 0.5 MPa average mechanical stress. The tendency of the disc to swell due to an osmotic potential was simulated by the addition of a 0.2 MPa pressureat the bony endplate and the annulus. The loading cycle was discretized into fixed time steps.

Denoziere and Ku (2006) applied for physiologic loads a pre-compression of 720 N simulating the intervertebral pressure of standing position and additional compressive forces were applied to the suitable areas of the endplates to simulate severe motions: 2000 N for lifting a load with straight legs in full flexion; 1000 N for full extension; 1300 N for full lateral bending; 11.45 Nm axial torque for axial rotation, based on the work of Nachemson (1966), White and Panjabi (1990) and Wilke et al. (1999). Zhong et al. (2006) analyzed a new cage by topology optimization by applying the maximum possible load without causing spinal injury, thus, 10 Nm for flexion, extension and torsion and lateral bending with 150 N compressive preload.

By analyzing rigid and dynamic fixation Rohlmann et al. (2007), vertebro- and kypho-plasty Rohlmann et al. (2006a) and dynamic implant Zander et al. (2006) applied follower load. An upper body weight of 260 N and a compressive follower load of 200 N were considered standing, flexion and extension. The follower load simulated the stabilizing effect of the local muscle forces. The lever arms to the disc center of erector spinae and rectus abdominis were 40 and 153 mm, respectively, and the direction of the muscle forces was quasi parallel to the lumbar spine. For axial rotation, a 10^0 axial pre-rotation was applied, and a follower load of 500 N representing the upper body weight and the muscle forces together was considered. Spinal loads were assumed to be 25% higher for walking than for standing. The method to estimate the muscle forces for standing and different inclination of the spine in the sagittal plane has been described in details by Zander et al. (2001), Wilke et al. (2003) and Rohlmann et al. (2006b).

Schmidt et al. (2007) in a degeneration analysis used unconstrained moment load 7.5 Nm with changing loading directions between each pair of main anatomical planes to simulate the combinations of the anatomical loadings, for example rotation with lateral bending, and so on. All these load cases were additionally combined with an axial compressive preload of 500 N. Interbody fusion and fixation techniques were compared by Chen et al. (2008) by applying a compressive preload of 150 N together with four different kinds of 10 Nm moments simulating the physiologic loading cases. Zhang et al. (2009) evaluated the load transfer of a dynamic stabilization device under compression, by applying axial compressive force of 2000 N for validating the model and 1000 N to investigate the load transmission characteristics of different implants.

Schmidt at al. (2009) analyzed flexible lumbar stabilization system by applying pure unconstrained moments in the three anatomical main planes, simulating flexion, extension lateral bending and axial rotation. The loads were increased incrementally in 10 load steps from zero to the predeterminated maximum values of 7.5 and 20 Nm. For poroelastic disc analysis for physiologic loading Malandrino et al. (2009) used 1000 N compression force load and 7.5 Nm moment load, similarly to Noailly et al. (2007).

By analyzing the effect of degenerative disc disease on adjacent segments Ruberté et al. (2009) applied different moment loads of flexion, extension, axial rotation and lateral bending for healthy (8, 6, 4 and 6 Nm, respectively) for healthy model, and these moments were modified for mild and moderate degenerations,. A compressive follower preload of 800 N was also applied.

Poroelastic creep analysis of a lumbar segment was investigated by Argoubi and Shirazi-Adl (1996). Th creep response of the segment was studied for a period of 2 hours under a constant axial force of 400, 1200 and 2000 N. Viscoelastic analysis of segment was investigated by Wang et al. (2000) for combined compression and sagittal flexion. A 600 N axial compressive load with a 60 N anterior shear load for a duration of 30 seconds simulated the preload of physiologic neutral posture. The hybrid loading mechanism was used to simulate the lowering task by specifying the final net compression of 2000 N, the anterior shear force of 200 N, and the sagittal flexion of 10^0. The three different durations of 0.3, 1 and 3 seconds represented the fast, medium and slow movements, respectively.

Dynamic loading of the L4-5 segment with poroelastic disc was analyzed by Williams et al. (2007) by applying short-term creep and cyclic loading. In simulating the short-term creep and standing recovery, a 400 N compressive preload was applied on the superior surface of L4, followed by an additional 400 N for 20 minutes, after which the load was reduced to 400N for 10 minutes for recovery. In simulating the short-term cyclic load and standing recovery, after the 400 N compressive preload, a peak-to-peak compressive force of 400 n and a peak-to-peak flexion moment of 5 Nm at a rate of 12 lifts per minute was applied for 20 minutes, then the disc was allowed to relax for 10 minutes with just the preload present.

Moramarco et al. (2010) validated the FE model of a lumbosacral segment by applying four simulation by different loading cases. First, an incremental 4 Nm pure flexion moment was applied. Then an axial compressive pre-loading of 100 N was applied first with a flexion moment than with an extension moment, both of 10Nm, applied incrementally. Finally, 10 Nm lateral moment was considered.

Kurutz and Oroszváry (2010) analyzed by FE simulation the stretching effect of a special underwater traction treatment when the patients are suspended cervically in vertical position in the water, supported on a cervical collar alone, loaded by extra lead weights on the ankles. There were two parts of the applied traction load: the removal of the compressive preload of body weight and muscle forces in water, named indirect traction load; and the direct traction load consisting of the tensile force of buoyancy with the applied extra loads. Based on mechanical calculations, for the standard body weight of 700 N, and the applied extra lead weights 40 N, the indirect and direct traction loads yields 840 *N* and 50 *N*, respectively.

9. Validation of the finite element models

By using FE models in a numerical simulation, the results should be trustworthy. But how do we know that the results or predictions can be believed with confidence? FE model validation can answer these questions. Correlation between FE results and experimental results can lead to use the FE model predictions with confidence. Recently there are many technologies for evaluating and improving the accuracy and validity of linear and nonlinear FE models.

In numerical simulations of biomechanics, for FE prediction accuracy assessment, a gold standard can be the *experimental validation of numerical results*. This enables the analyst to improve the quality and reliability of the FE model and the modeling methodology. If there is very poor agreement between the analytical and experimental data, by using certain nu-

merical techniques for model updating allow the user to create improved models which represent reality much better than the original ones.
For example, analyzing the effect of disc height Lu et al. (1996) validated the FE model by direct comparison of the model predictions with experimental results of axial displacement, axial compressive stress and posterolateral disc bulge obtained on cadaveric motion segments. In viscoelastic analysis Wang et al. (2000) validated the segment model by the experimental data of constant compressive strain rate loading, creep loading and cyclic relaxation loading. In analyzing the adjacent segment syndrome Chen et al. (2001) validated the L1-5 multisegment model by comparing the kinematics data of the model with in vivo experiments under the same loading condition. In fluid flow analysis within the disc Ferguson et al. (2004) validated the FE model against in vitro creep/swelling data for isolated discs. Denoziere and Ku (2006) validated the segment model by comparing the average mobility of the healthy model in flexion-extension, lateral bending and axial rotation with various experimental reports on vertebral motions. In short-term creep and cyclic analysis Williams et al. (2007) validated the model by in vivo creep and recovery disc height variations. For interbody fusion and fixation analysis Chen et al. (2008) validated the intact FE model by comparing the flexion-extension angles with experimental data. Moramarco et al. (2010) validated the FE model of a lumbosacral segment by in vitro experiments of axial displacements and posterior disc bulge. Kurutz and Oroszváry (2010) validated the lumbar segment model for both compression and tension and for both healthy and degenerated disc. Distribution of vertical compressive stresses of healthy and degenerated discs in the mid-sagittal horizontal section of the disc was compared with the experimental results of Adams et al. (1996, 2002), obtained by stress profilometry. In axial tension, the calculated disc elongations were compared with the in vivo measured elongations of Kurutz (2006a) for healthy and degenerated segments.

10. Types of elements applied to the segmental structures

The cancellous core and the posterior bony elements of vertebrae can be modeled as *3D solid continuum elements*, as isoparametric 8-node hexahedral (brick) elements, or as 20- or 27-noded brick elements, moreover, as 10-noded tetrahedral elements. The cortical shell and the endplates can be modeled as *thin shell elements*, like 4-node shell elements. Quasi-rigid *beam elements* can connect the posterior vertebra with the medial transverse processes (pedicles) and from the medial transverse processes to the medial spinous process (lamina). Beam elements can also be used to represent the transverse and spinous processes. The bony surface of the facet joints can be represented by shell elements where beam elements link these facets to the lamina, simulating the inferior and superior articular processes (Little et al. 2008). The facet joints can be modeled as 3D 8-node surface-to-surface *contact elements* (Zhong et al., 2006).
The disc annulus ground substance is generally modeled as 3D continuum elements. The collagen fibers can be modeled as *truss elements* or as *reinforced bar (rebar) type elements* embedded in 3D solid elements (Lodygowski et al. 2005). The nucleus pulposus can be modeled as *hydrostatic fluid volume elements*.
The anterior and posterior longitudinal ligaments can be modeled as thin shell elements, or, the ligaments can be modeled as 2-node axial elements, that is, *tension only linear or nonlinear truss or cable or spring elements*.

11. Example: FE numerical simulation of age-related degeneration of lumbar segments

Spinal aging degeneration processes are the most typical example of those phenomena that can not be clarified in their progress by experimental methods, but exclusively by numerical simulation. 3D FE simulation of age-related degeneration processes of lumbar segments L3-S1 was investigated in axial compression. Aging degeneration of the segment was modeled by the material properties of ist components, validated both for compression and tension, by comparing the numerical results with experimental data. Five grades of aging degeneration were distinguished from the healthy to fully degenerated case.

A 3D geometrical model of a typical lumbar FSU was created, obtained by using Pro/Engineer code. The geometrical data of the FSU were obtained by the measures of a typical lumbar segment. Cortical and cancellous bones of vertebrae were separately modeled, including posterior bony elements, too. The thickness of vertebral cortical walls and endplates were 0.35 and 0.5 mm, respectively. The height of the disc was 10 mm. Annulus fibrosus consisted of ground substance and elastic fibers. Annulus matrix was divided to internal and external ring; with three layers of annulus fibers of 0.1 mm^2 cross section. The geometry and orientation of facet joints were chosen according to Panjabi et al., (1993).

The FE mesh was generated by ANSYS Workbench, the connections between several geometrical components were integrated to the FE model by ANSYS Classic. The FE model consisted of solid, shell and bar elements. Annulus matrix, nucleus, cancellous bone, articular joints and different types of attachments were modeled by Solid_186/187 elements with quadratic displacement behavior. Cortical shells and endplates were modeled by Shell_181 elements with four nodes at each element. All ligaments were modeled by Shell_41 elements, with tension-only material. Annulus fibers were mapped into Link_10 bar elements with bilinear stiffness matrix resulting in a uniaxial tension-only behaviour.

By means of a systematic numerical analysis of the separated effect of the two main mechanical components of aging degeneration, it was proved that at the beginning period of the aging process, the effect of the loss of hydrostatic stress state of the nucleus had the dominant effect, while in further aging, the stiffening of nucleus dominated. This fact leads to the largest deformability and the smallest compressive stiffness, consequently, to the risk of segmental instability at mildly degenerated case in young age, while the stiffness and stability increased with further aging and degeneration. Exclusively by FE numerical analysis the observation and question of many international papers, why low back pain problems insult so frequently the young adults, could be answered.

For the introduced numerical model, the mean axial compressive stiffness of the nucleus, internal and external annulus was about 700, 1200, 500 N/mm for healthy; 500, 1000, 400 N/mm for mildly; 800, 1200, 600 N/mm for medium; 2100, 2000, 1100 N/mm for severely and 5500, 4500, 3000 N/mm for fully degenerated discs, respectively. The stiffness of the whole disc was about 2400, 1900, 2600, 5200 and 13000 N/mm fort he five degeneration grades. While vertical intradiscal stresses showed significant change during aging degeneration, between 0.6-1.6 MPa, the horizontal stresses remained quasi constant and small, between 0.2-0.5 MPa for 1000 N compression. In the numerical modeling of hydrostatic state of a healthy nucleus, smaller than 1 MPa Young's modulus of nucleus must be considered to cut down the nuclear stress divergence below 10%. For healthy nucleus, E=0.1 MPa seemed to be acceptable.

FE simulations of degeneration processes of lumbar segments may help clinicians to understand the initiation and progression of disc degeneration and to treat lumbar discopathy problems even more effectively.

12. References

Acaroglu, E.R., Iatridis, J.C., Setton, L.A., Foster, R.J., Mow, V.C., Weidenbaum, M. (1995). Degeneration and aging affect the tensile behaviour of human lumbar anulus fibrosus, *Spine*, 20(24), 2690-2701.

Argoubi, M., Shirazi-Adl, A. (1996). Poroelastic creep response analysis of a lumbar motion segment in compression, *Journal of Biomechanics,* 29(10), 1331-1339.

Adams, M.A., Dolan, P. (1991). A technique for quantifying the bending moment acting on the lumbar spine in vivo, *Journal of Biomechanics,* 24(2), 117-126.

Adams, M.A., Freeman, B.J., Morrison, H.P., Nelson, I.W., Dolan, P. (2000), Mechanical initiation of intervertebral disc degeneration, *Spine,* 25(13), 1625-1636.

Adams, M.A., Bogduk, N., Burton, K., Dolan, P. (2002). *The Biomechanics of Back Pain*, Churchill Livingstone, Edinburgh, London, New York.

Adams, M.A., Hutton, W.C. (1983). The effect of posture on the fluid content of lumbar intervertebral discs, *Spine,* 8(6), 665-671.

Adams, M.A., McNally, D.S., Dolan, P. (1996). Stress distributions inside intervertebral discs. The effects of age and degeneration. *J. Bone Joint Surg. Br.* 78(6), 965-972.

Bader, D.L., Bouten, C. (2000). Biomechanics of soft tissues. In: Dvir, Z. (Ed.), *Clinical Biomechanics*, Churchill Livingstone, New York, Edinburgh, London, Philadelphia, pp. 35-64.

Banse, X., Devogelaer, J.P., Munting, E., Delloye, C., Cornu, O., Grynpas, M. (2001). Inhomogeneity of human vertebral cancellous bone: systematic density and structure patterns inside the vertebral body. *Bone*, 28(5), 563-571.

Baroud, G., Nemes, J., Heini, P., Steffen, T. (2003). Load shift of the intervertebral disc after a vertebroplasty: a finite element study, *European Spine Jourbal,* 12(4), 421-426.

Benzel, E.C.: *Biomechanics of Spine Stabilization,* (2001). American Association of Neurological Surgeons, Rolling Meadows, Illinois.

Bogduk, N. (2005), *Clinical Anatomy of the Lumbar Spine and Sacrum,* Elsevier, Churchill Livingstone, Edinburgh, London, New York, Oxford, Philadelphia, St Luis, Sidney, Toronto.

Bogduk, N., Twomey, L.T. (1987). *Clinical Anatomy of the Lumbar Spine,* Churchill Livingstone, New York.

Briggs, A.M., Greig, A.M., Wark, J.D., Fazzalari, N.L., Bennell, K.L. (2004). A review of anatomical and mechanical factors affecting vertebral integrity. *International Journal of Medical Sciences,* 1, 170-180.

Cassidy, J.J., Hiltner, A., Baer, E. (1989). Hierarchical structure of the intervertebral disc, *Connective Tissue Research*, 23(1), 75-88.

Cassinelli, E., Kang, J.D. (2000). Current understanding of lumbar disc degeneration, *Operative techniques in orthopaedics*, 10(4), 254-262.

Chen, C.S., Cheng, C.K., Liu, C.L., Lo, W.H. (2001). Stress analysis of the disc adjacent to interbody fusion in lumbar spine, *Medical Engineering and Physisc,* 23(7), 483-491.

Chen, S.H., Tai, C.L., Lin, C.Y., Hsieh, P.H., Chen, W.P. (2008). Biomechanical comparison of a new stand-alone anterior lumbar interbody fusion cage with established fixation techniques – a three-dimensional finite element analysis, *BMC Muscoskeletal Disorders*, 9(88), 1-10.

Denoziere, G., Ku, D.N. (2006). Biomechanical comparison between fusion of two vertebrae and implantation of an artificial intervertebral disc, *Journal of Biomechanics*, 39(4), 766-775.

Dolan, P., Mannion, A.F., Adams. M.A. (1994). Passive tissues help the back muscles to generate extensor moments during lifting, *Journal of Biomechanics,* 27(8), 1077-1087.

Dolan, P., Adams, M.A. (2001). Recent advances in lumbar spinal mechanics and their significance for modelling, *Clin. Biomech., 16(Suppl.), S8-S16, 2001.*

Duan, Y., Seeman, E., Turner, C.H. (2001). The biomechanical basis of vertebral body fragility in men and women, *Journal of Bone and Mineral Research*, 16(12), 2276-2283.

Dvir, Z. (2000). *Clinical Biomechanics,* Churchill Livingstone, New York, Edinburgh, London, Philadelphia.

Eberlein, R., Holzapfel, G.A., Fröhlich, M. (2004). Multi-segment FEA of the human lumbar spine including the heterogeneity of the annulus fibrosus, *Computational Mechanics*, 34(2), 147-163.

Elliott, D.M., Setton, L.A. (2001). Anisotrope and inhomodeneous tensile behaviour of the human annulus fibrosus: experimental measurements and material model predictions, *Journal of Biomechanical Engineering*, 123, 256-263.

Fagan, M.J., Julian, S., Siddall, D.J., Mohsen, A.M. (2002). Patient specific spine models. Part 1: finite element analysis of the lumbar intervertebral disc – a material sensitivity study, *Proceedings of the Institution of Mechanical Engineers*, Part H: 216(5), 299-314.

Ferguson, S.J., Steffen, T. (2003). Biomechanics of the aging spine, *European Spine Journal*, Suppl 2, S97-S103.

Ferguson, J., Ito, K., Nolte, L.P. (2004). Fluid flow and convective transport of solutes within the intervertebral disc, *Journal of Biomechanics*, 37(2), 213-221.

Goel, V.K., Kong, W., Han, J.S., Weinstein, J.N., Gilbertson, L.G. (1993). A combineg finite elelent and optimization investigation of lumbar spine mechanics with and without muscles. *Spine*, 18, 1531-1541.

Goel, V.K., Monroe, B.T., Gilbertson, L.G., Brinckmann, P. (1995a). Interlaminar shear stresses and laminae separation in the disc. Finite element analysis of the L3-L4 motion segment subjected to axial compressive loads. *Spine*, 20(6), 689-698.

Goel, V.K., Gilbertson, L.G. (1995b). Applications of the finite element method to thoracolumbar spinal research – past, present and future, *Spine,* 20(15), 1719-1727.

Gong, H., Zhang, M., Yeung, H.Y., Qin, L. (2005). Regional variations in microstructural properties of vertebral trabeculae with aging. *Journal of Bone and Mineral Metabolism*, 23(2), 174-180.

Goto, K., Tajima, N., Chosa, E., Totoribe, K., Kuroki, H., Arizumi, Y. (2002). Mechanical analysis of the lumbar vertebrae in a three-dimensional finite element method in which intradiscal pressure int he nucleus pulposus was used to establish the model, *Journal of Orthopedic Science*, 7(2), 243-246.

Gu, W.Y., Mao, X.G., Foster, R.J., et al., (1999). The anisotropic hydraulic permeability of human lumbar annulus fibrosus – influence of age, degeneration, direction and water content, *Spine*, 24(23), 2449-2455.

Iatridis, J.C., Setton, L.A., Wedenbaum, M., Mow, V.C. (1997). Alterations in the mechanical behavior of the human lumbar nucleus pulposus with degeneration and aging. *Journal of Orthopaedic Research*, 15(2), 318-322.

Iatridis, J.C., Setton, L.A., Foster, R.J., Rawlins, B.A., Weidenbaum, M., Mow, V.C. (1998). Degeneration affects the anisotropic and nonlinear behaviours of human anulus fibrosus in compression, *Journal of Biomechanics*, 31(6), 535-544.

Johannessen, W., Elliott, D.M. (2005). Effects of degeneration on the biphasic material properties of human nucleus pulposus in confined compression, *Spine*, 30(24), E724-E729.

Keaveny, T.M., Yeh, O.C. (2002). Architecture and trabecular bone - toward an improved understanding of the biomechanical effects of age, sex and osteoporosis. *Journal of Musculoskeletal and Neuronal Interactions*, 2(3), 205-208.

Kim, Y.E., Goel, V.K., Weinstein, J.N., Lim, T.H. (1991). Effect of disc degeneration at one level on the adjacent level in axial mode, *Spine*, 16(3), 331-335.

Kumaresan, S., Yoganandan, N., Pintar, F.A. (1999). Finite element analysis of the cervical spine: a material property sensitivity study, *Clinical Biomechanics*, 14(1), 41-53.

Kumaresan, S., Yoganandan, N., Pintar, F.A. et al. (2001). Contribution of disc degeneration to osteophyte formation in the cervical spine: a biomechanical investigation, *Journal of Orthopaedic Research*, 19(5), 977-984.

Kurowski, P., Kubo, A. (1986). The relationship of degeneration of the intervertebral disc to mechanical loading conditions on lumbar vertebrae, *Spine*, 11(7), 726-731.

Kurutz, M., (2006a). Age-sensitivity of time-related in vivo deformability of human lumbar motion segments and discs in pure centric tension, *Journal of Biomechanics, 39*(1), 147-157.

Kurutz, M., (2006b). In vivo age- and sex-related creep of human lumbar motion segments and discs in pure centric tension, *Journal of Biomechanics, 39*(7), 1180-9.

Kurutz, M., Bene É., Lovas, A., (2003). In vivo deformability of human lumbar spine segments in pure centric tension, measured during traction bath therapy, *Acta of Bioengineering and Biomechanics, 5*(1), 67-92.

Kurutz, M., Oroszváry, L. (2010). Finite element analysis of weightbath hydrotraction treatment of degenerated lumbar spine segments in elastic phase, *Journal of Biomechanics,* 43(3), 433-441.

Lavaste, F., Skalli, W., Robin, S., Roy-Camille, R., Mazel, C. (1992). Three-dimensional geometrical and mechanical modelling of lumbar spine, *Journal of Biomechanics, 25*(10), 1153-1164.

Little, J.P., Pearcy, M.J., Adam, C.J. (2008). Are coupled rotations in the lumbar spine largely due to the osseo-ligamentous anatomy? - a modeling study. *Computer Methods in Biomechanics and Biomedical Engineering,* 11(1), 95-103.

Lodygowski, T., Kakol. W., Wierszycki, M. (2005). Three-dimensional nonlinear finite element model of lumbar intervertebral disc, Acta of Bioengineering and Biomechanics, 7(2), 29-37.

Lu, Y.M., Hutton, W.C., Gharpuray, V.M. (1996a). Can variation in Intervertebral disc height affect the mechanical function of the disc? *Spine*, 21(19), 2208-2217.

Lu, Y.M., Hutton, W.C., Gharpuray, V.M. (1996b). Do bending, twisting and diurnal fluid changes in the disc affect the propensity to prolapse? A viscoelastic finite elelent model. *Spine*, 21(22), 2570-2579.

Malandrino, A., Planell, J.A., Lacroix, D., (2009). Statistical factorial analysis ont he poroelastic material properties sensitivity of the lumbar intervertebral disc under compression, flexion and axial rotation, *Journal of Biomechanics*, 42(3), 341-348.

McGill, S.M. (200). Biomechanics of the thoracolumbar spine, In: Dvir, Z. (Ed.), *Clinical Biomechanics*, Churchill Livingstone, New York, Edinburgh, London, Philadelphia, pp. 103-139.

McNally, D.S., Adams, M.A. (1992). Internal intervertebral disc mechanics as revealed by sress profilometry, *Spine*, 17(1), 66-73.

Moramarco, V., Palomar, A.P., Pappalettere, C., Doblaré, M. (2010). An accurate validation of a computational model of human lumbosacral segment, *Journal of Biomechanics*, 43(2), 334-342.

Mosekilde, L. (2000). Age-related changes in bone mass, structure, and strength effects of loading, *Zeitschrift für Rheumatologie*, 59(Suppl.1), 1-9.

Nachemson, A. (1966). The load on lumbar discs in different positions of the body, Clinical *Orthopaedics and Related Research*, 45, 107-122.

Nachemson, A.L. (1981). Disc pressure measurements, *Spine,* 6(1), 93-97.

Nachemson, A.L., Morris, J.M. (1964). In vivo measurements of intradiscal pressure, *Journal of Bone and Joint Surgery Am*., 46, 1077-1092.

Natarajan, R.N., Williams, J.R., Andersson, G.B. (2004). Recent advances in analytical modelling of lumbar disc degeneration. *Spine*, 29(23), 2733-2741.

Natarajan, R.N., Williams, J.R., Andersson, G.B., 2006. Modeling changes in intervertebral disc mechanics with degeneration, *Journal of Bone and Joint Surgery Am*. 88(4), Suppl. 2, 36-40.

Noailly, J., Lacroix, D., Planell, J.A. (2005). Finite element study of a novel intervertebral disc substitute, *Spine*, 30(20), 2257-2264.

Noailly, J., Wilke, H.J., Planell, J.A., Lacroix, D. (2007). How does the geometry affect the internal biomechanics of a lumbar spine bi-segment finite element model? Consequences on the validation process, *Journal of Biomechanics*, 40(11), 2780-2788.

Panjabi, M.M., Oxland, T., Takata, K., Goel, V., Duranceau, J., Krag, M., (1993). Articular facets of the human spine, quantitative three dimensional anatomy, *Spine*, 18, 1298-1310.

Pintar, F.A., Yoganandan, N., Myers, T. Elhagediab, A., Sances Jr, A. (1992). Biomechanical properties of human lumbar spine ligaments, *Journal of Biomechanics*, 25(11), 1351-1356.

Rohlmann, A., Zander, T., Bergmann, G. (2006a). Spinal loads after osteoporotic vertebral fractures treatedby vertebroplasty or kyphoplasty, *European Spine Journal*, 15(8), 1255-1264.

Rohlmann, A., Bauer, L., Zander, T., Bergmann, G., Wilke, H.J., (2006b). Determination of trunk muscle forces for flexion and extension by using a validated finite element model of the lumbar spine and measured in vivo data, *Journal of Biomechanics*, 39(6), 981-989.

Rohlmann, A., Zander, T., Schmidt, H., Wilke, H.J., Bergmenn, G. (2006c). Analysis of the influence of disc degeneration on the mechanical behaviour of a lumbar motion segment using the finite element method, *Journal of Biomechanics*, 39(13), 2484-2490.

Rohlmann, A., Burra, N.K., Zander, T., Bergmann, G. (2007). Comparison of the effect of bilateral posterior dynamic and rigid fixation devices ont he loads in the lumbar spine: a finite element analysis, *European Spine Journal*. 16(8), 1223-1231.

Ruberté, L.M., Natarajan, R.N., Andersson, G.B. (2009). Influence of single-level lumbar degenerative disc disease on the behavior of the adjacent segments--a finite element model study, *Journal of Biomechanics*, 42(3), 341-348.

Sato, K., Kikuchi, S., Yonezawa, T. (1999). In vivo intradiscal pressure measurement in healthy individuals and in patients with ongoing back problems, *Spine*, 24(23), 2468-2474.

Schmidt, H., Heuer, F., Wilke, H.J. (2009). Which axial and bending stiffnesses of posterior implants are required to design a flexible lumbar stabilization system? *Journal of Biomechanics*, 42(1), 48-54.

Schroeder, Y., Wilson, W., Huyghe, J.M., Baaijens, P.T. (2006). Osmoviscoelastic finite element mdel of the intervertebral disc, *European Spine Journal*, 15(Suppl 3), 361–371.

Sharma, M., Langrana, M.A., Rodriguez, J. (1995). Role of ligaments and facets in lumbar spinal stability, *Spine*, 20(8), 887-900.

Shirazi-Adl, S.A., Shrivastava, S.C., Ahmed, A.M., (1984). Stress analysis of the lumbar disc-body unit in compression. A three-dimensional nonlinear finite element study, *Spine*, 9(2), 120-34.

Shirazi-Adl, S.A., Ahmed, A.M., Shrivastava, S.C., (1986a). Mechanical response of a lumbar motion segment in axial torque alone and combined with compression, *Spine*, 11(9), 914-927.

Shirazi-Adl, A., Ahmed, A.M., Shrivastava, S.C., (1986b). A finite element study of a lumbar motion segment subjected to pure sagittal plane moments, *Journal of Biomechanics*, 19(4), 331-350.

Smit, T., Dodgaard, A., Schneider, E. (1997). Structure and function of vertebral trabecular bone, *Spine*, 22(24), 2823-2833.

Umehara, S., Tadano, S., Abumi, K. (1996). Effects of degeneration on the elastic moduls distribution in the lumbar intervertebral disc, *Spine*, 21(7), 811-819.

Wang, J.L., Parnianpour, M., Shirazi-Adl, A., Engin, A.E. (2000). Viscoelastic finite element analysis of a lumbar motion segment in combined compression and sagittal flexion, *Spine*, 25(3), 310-318.

White, A. A., Panjabi, M. M. (1990). *Clinical Biomechanics of the Spine*, Lippincott Williams and Wilkins, Philadelphia, etc.

Wilke, H.J., Neef, P., Caimi, M., Hoogland, T., Claes, L.E. (1999). New in vivo measurements of pressures in the intervertebral disc in daily life, *Spine*, 24(8):755-62.

Wilke, H.J., Rohlmann, A., Neller, S., Graichen, F., Claes, L., Bergmann, G. (203). ISSLS Prize Winner: A novel approach to determine trunk muscle forces during flexion and extension: a comparison of data from an in vitro experiment and in vivo measurements, *Spine*, 28(23), 2585-2593.

Williams, J.R., Natarajan, R.N., Andersson, G.B.J. (2007). Inclusion of regional poroelastic material properties better predicts biomechanical behaviour of lumbar discs subjected to dynamic loading, *Journal of Biomechanics*, 40(9), 1981-1987.

Zander, T., Rohlmann, A., Calisse, J., Bergmann, G., (2001). Estimation of muscle forces in the lumbar spine during upper body inclination, *Clinical Biomechanics*, 16(Suppl1), S73-S80.

Zander, T., Rohlmann, A., Burra, N.K., Bergmann, G., (2006). Effect of a posterior dynamic implant adjacent to a rigid spinal fixator, *Clinical Biomechanics*, 21(8), 767-774.

Zhang, Q.H., Zhou, Y.L., Petit, D., Teo, E.C., (2009). Evaluation of load transfer characteristics of a dynamic stabilization device on disc loading under compression, *Medical Engineering and Physics*, 31(5), 533-538.

Zhong, Z.C., Wei, S.H., Wang, J.P., Feng, C.K., Chen, C.S., Yu, C.H. (2006). Finite element analysis of the lumbar spine with a new cage using a topology optimization method, *Medical Engineering and Physics*, 28(1), 90-98.

IntechOpen

IntechOpen

Finite Element Analysis
Edited by David Moratal

ISBN 978-953-307-123-7
Hard cover, 688 pages
Publisher Sciyo
Published online 17, August, 2010
Published in print edition August, 2010

Finite element analysis is an engineering method for the numerical analysis of complex structures. This book provides a bird's eye view on this very broad matter through 27 original and innovative research studies exhibiting various investigation directions. Through its chapters the reader will have access to works related to Biomedical Engineering, Materials Engineering, Process Analysis and Civil Engineering. The text is addressed not only to researchers, but also to professional engineers, engineering lecturers and students seeking to gain a better understanding of where Finite Element Analysis stands today.

How to reference
In order to correctly reference this scholarly work, feel free to copy and paste the following:

Marta Kurutz (2010). Finite Element Modelling of Human Lumbar Spine, Finite Element Analysis, David Moratal (Ed.), ISBN: 978-953-307-123-7, InTech, Available from: http://www.intechopen.com/books/finite-element-analysis/finite-element-modelling-of-human-lumbar-spine-

INTECH
open science | open minds

InTech Europe
University Campus STeP Ri
Slavka Krautzeka 83/A
51000 Rijeka, Croatia
Phone: +385 (51) 770 447
Fax: +385 (51) 686 166
www.intechopen.com

InTech China
Unit 405, Office Block, Hotel Equatorial Shanghai
No.65, Yan An Road (West), Shanghai, 200040, China
中国上海市延安西路65号上海国际贵都大饭店办公楼405单元
Phone: +86-21-62489820
Fax: +86-21-62489821

© 2010 The Author(s). Licensee IntechOpen. This chapter is distributed under the terms of the Creative Commons Attribution-NonCommercial-ShareAlike-3.0 License, which permits use, distribution and reproduction for non-commercial purposes, provided the original is properly cited and derivative works building on this content are distributed under the same license.