Acta of Bioengineering and Biomechanics
Vol. 14, No. 1, 2012

Original paper

Finite element analysis of artificial disc with an elastomeric core in the lumbar spine

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This paper presents the application of finite element method in an artificial disc modelling. The prosthesis consisted of two metal plates and a flexible elastomeric core made of the nanocomposite polyurethane. Two types of connections between the plates and the core were compared: the device with an integral inlay and the device with a separate inlay coming into contact with the plates. The artificial disc with a separate inlay imitated better the human intervertebral disc. The main target of this paper was to evaluate the characteristics of force–displacement and moment–angle for the new design of the prosthesis with a separate inlay under compression, sagittal bending, shear and axial rotation. For some analyzed cases except the axial rotation and shear, where the prosthesis was too flexible, the results were roughly similar to those observed in the human spinal segment. The material effort in the prosthesis under compressive load was comparable in both types of connections between the plates and the core.

Key words: *FE analysis, artificial disc, elastomeric core, lumbar spine*

1. Introduction

Nowadays many people suffer from spinal disorders which may be a cause of the low back pain syndrome. One of the treatment methods is replacing the degenerative patient's disc with an artificial device such as a prosthesis which relieves the pain and restores physiological motion of the operated spinal segment, consisting of two vertebras and an intervertebral disc [1]. The prosthesis should work safely for about fifty years and should not cause allergic response of the human body. The components of the prosthesis have to satisfy many criteria. One of them is to ensure low stress and strain levels within the ranges of loads and motions that occur in the spine. Stress and strain evaluation in the spinal segment with the artificial disc is a complex problem and thus involves numerical simulation. This paper presents the application of the finite element method in an artificial disc modelling.

Two types of artificial discs can be identified: the ball-joint and flexible prostheses. The first type consists of two rigid metal plates with a convex inlay [2] or two rigid metal components (convexo-concave) [3]. Currently, the ball-joint devices are most commonly implanted. Recently, some results were published indicating that inaccurate imitation of kinematics and insufficient resistance to the shearing forces in the spinal segments with the ball-joint prostheses were the causes of injuries to the facet joints [4].

As quite distinct from this design, the flexible device has an elastomeric part imitating the human disc. The main problem is how to fix the inlay to the vertebrae, so few devices were implanted. They additionally had two porous metal plates adjacent to

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Received: September 23rd, 2010
Accepted for publication: January 18th, 2012

the inlay which enabled the prosthesis to grow into the vertebral body [4], [5]. The second question is how to avoid the fatigue of the inlay which causes fracture propagation inside the elastomeric inlay or at the interface between the inlay and plates [5]. This is particularly important if the interface is inseparable. Therefore, the shape of an outer region of the elastomeric inlay should be specially designed to prevent relatively high strains occurring during translational and rotational movements. The material properties of the inlay should be fitted to the load magnitudes and the ranges of motion occurring in the spine. Additionally the joint between the inlay and the plates should be resistant to wear. These requirements could be met due to a choice of a composite material whose properties could be adjusted and even changed across the inlay. For instance, nanocomposite polyurethane exhibits features like biocompatibility, strength and fatigue resistance, depending on its chemical composition, and reveals good cohesion at the interface between metallic alloys. The new manufacturing technology of this material was recently developed at WUT [6]. Hence, the nanocomposite polyurethane is considered as the material which could be applied in an elastomeric inlay [4].

Another design of the flexible artificial disc is the prosthesis with a frictional joint, where the separate elastomeric inlay could slip between the plates. The main drawback of this device is obviously wear debris, but the use of a more rigid and resistant material of the inlay could result in similar flexibility as that achieved for the prosthesis with an integral inlay. This type of joint could also be characterized by the considerably lower strain and stress values in comparison with those of an integral inlay, especially during rotational movements. Besides, the common problem with the ball-joint and flexible devices lay in a fusion of the implanted spinal segment, caused probably by the implant design and (or) the method of implantation (to what extent the degenerative disc was removed). This fusion was created by tissue ingrowth between the implant plates. Hence, for example, it seems that short-term successful results of implantation of the modern flexible prosthesis *Physio-L* [4] should be verified in a longer time period because of the possibility of spinal segment fusion. In the case of the prosthesis with a separate inlay, the outer regions of the elastomeric inlay need not to be removed because of high strain, so it seems to be less susceptible to unwanted fusion. Thus, this type of the elastomeric prosthesis was mainly taken into account in this paper.

2. Methods

2.1. Finite element model

The finite element model of the functional spinal segment included the fragments of lumbar vertebral bodies similar to L4 and L5 and the prosthesis. Both types of the flexible artificial disc were analyzed (figure 1). The prostheses consisted of two metal plates and the elastomeric inlay. The size of intervertebral space was the same for the prosthesis with an integral inlay and that with a separate inlay. The isotropic linear model of material was assumed for the cortical bone, trabecular bone and the metal plate made of Co–Cr–Mo alloy (the table) [7]. A nonlinear hiperelastic incompressible Gent material was used for the inlay model made of polyurethane nanocomposite (1):

$$W=\frac{1}{2}\left[G_0 J_m \ln\left\{1-\frac{(I_1-3)}{J_m}\right\}^{-1}+\frac{1}{d}(J^2-1-2\ln J)\right], \tag{1}$$

where:

W – the strain energy,

G_0 = 5 MPa – the initial shear modulus,

J_m = 80 – the limiting value of $(I_1–3)$; $I_1 = J^{-2/3}\cdot(\lambda_1^2 + \lambda_2^2 + \lambda_3^2)$,

J – the volume ratio,

λ_i – the principal extension ratios,

d = 0 – the incompressibility parameter, $d = 2/K$ (K – the bulk modulus).

Table. Isotropic material properties

Part	Young's modulus E (MPa)	Poisson's ratio ν
Cortical bone	12 000	0.35
Trabecular bone	400	0.40
Co–Cr–Mo plate	200 000	0.30

In both cases shown in figure 1, metal plates were bound to the vertebrae. The model analysed was symmetrical in respect of the horizontal and vertical planes. A load was applied as the displacements imposed on the upper vertebral body section. Except the compression case, the displacements were applied also to the lower vertebral body section in order to keep a constant value of the compressive force. The ranges of motion for load cases were taken from the literature [8]. Forces and moments were calculated based on the nodal forces at the transversal section

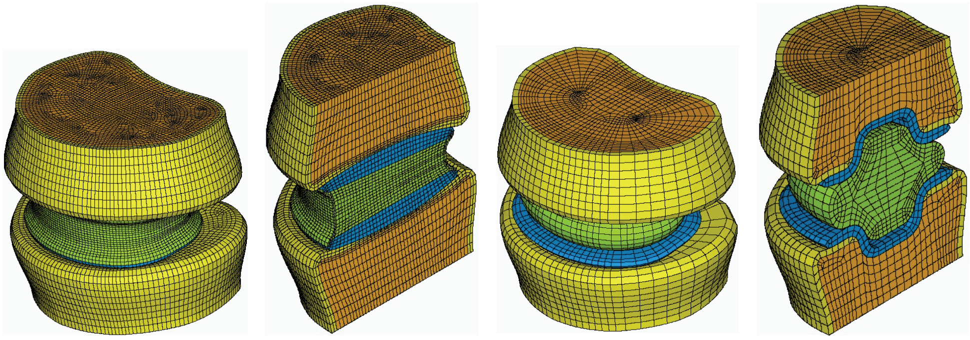

Fig. 1. FE model of the spinal unit with an integral inlay (left) and a separate inlay (right)

area of the inlay. In the model with a separate inlay, there was a contact with friction between the plates and the inlay ($\mu = 0.05$). The ANSYS system (v. 11) allowed us to evaluate the displacements, stress and strain components in bones and in the prosthesis. The structural solid Type 45 and Type 185 elements and the surface-to-surface contact elements, Type 174 and Type 170, were used.

2.2. Compression case

Since the primary load occurring in the lumbar spine was compression, we chose this case to compare two artificial discs: that with an integral inlay and that with a separate inlay. The aim of simulation was to evaluate the force–displacement characteristics of the prosthesis. Simulations were carried out for a neutral position of the spinal segment. The angle between vertebrae was 6°. The load was applied as the displacements imposed on the upper vertebral body section. Because of symmetry, one quarter of the full model of the prosthesis was analyzed and symmetry condition was applied to the transversal section area of the inlay.

2.3. Sagittal bending, sagittal shear and axial rotation cases

Sagittal bending often occurs during daily activity. On the basis of literature the range of motion was assumed to be +13° for flexion and –7° for extension [8]. The simulation of sagittal bending was run for a full range of motion under constant compressive load of 1 kN (figure 6). Half-model of the prosthesis with a separate inlay was analyzed in respect of the symmetry plane. Load was applied as the imposed displacements which produced circumferential movements of the upper plate round the centre of rotation placed near the lower vertebral body surface between one third and one half of the disc length measured from the side of the vertebral canal [8]. The support onto the lower surface was adjusted to keep a constant value of the compressive force. The bending moment was calculated on the basis of nodal forces acting on the transversal section area of the inlay and their distances from the centre of rotation (figure 6). The starting angle between the vertebral body surfaces was 6° (neutral position of the spinal segment).

The sagittal shear case was important because of possible facet joint injury occurring in the spinal segments with ball-joint prostheses which were not able to resist shear loads [4]. During the simulation the upper plate of the prosthesis moved horizontally ±4 mm in sagittal plane relative to the lower plate under constant compressive load of 1 kN.

The axial rotation case had a similar importance to that of the sagittal shear. The range of the rotation angle was ±3.6° from the neutral position. The torsional moment was transferred by the prosthesis with a separate inlay because of the elliptical shape of the inlay protrusions.

3. Results

Both types of the flexible artificial discs were compared only for the compression case. In the prosthesis with an integral inlay (made of the same material as a separate inlay), large deformations occurred for the cases of sagittal bending and shear which caused high stress levels. So the results of the artificial disc with an integral inlay are presented only for compression (subsection 3.1). The results in subsections 3.2, 3.3 and 3.4 relate only to the prosthesis with a separate inlay.

3.1. Comparison of two types of prostheses under compression

Figures 2, 3 and 4 show the basic results obtained for both prostheses under the compressive force of 3 kN, which can be often experienced during daily life.

The force–displacement characteristics under compression for the two types of implants were shown in figure 5. The displacement *u* was measured as a decrease in the distance between vertebral body surfaces in the middle of the vertebrae. The device with an integral inlay was about four times stiffer than the prosthesis with a separate inlay (see also figure 2).

The approximate stiffness coefficients of the intact spinal segment come from the monograph written by WHITE and PANJABI [8]. More accurate data [9] indicate that within the range of compressive force from 0 to 1 kN the stiffness of an intervertebral human disc equals 0.8 kN/mm and increases to 2 kN/mm when the compressive forces exceed 4 kN. Thus the prosthesis with an integral inlay is too stiff in comparison with the intact human disc.

As quite distinct from this design, the stiffness of the prosthesis with a separate inlay was similar to that observed in the intervertebral human disc. If the displacement exceeded 2 mm, the gap between the frontal surface of the inlay protrusion and the internal surface of plate was closed, so the stiffness increased significantly in accordance with the experimental data cited above.

Stress distribution in the integral inlay was strongly non-uniform (figure 4). The highest value occurred in the region of the interface between the plate and the inlay in the rear part of the prosthesis. The equivalent stress in the inlay proved to be more uniform. Its highest value was observed in the outer

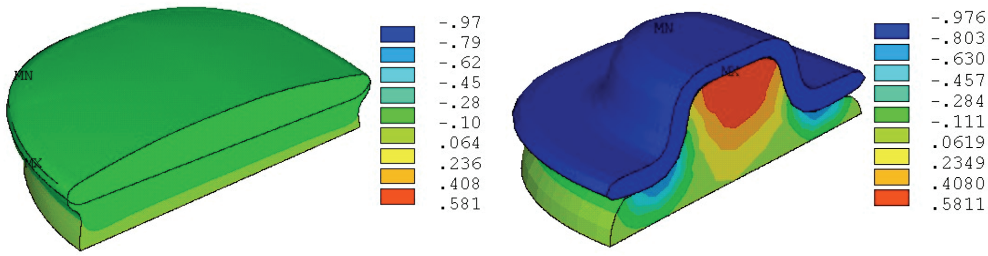

Fig. 2. Vertical displacement (mm) in the upper part of the inlay at compressive force of 3 kN: integral inlay (left), separate inlay (right)

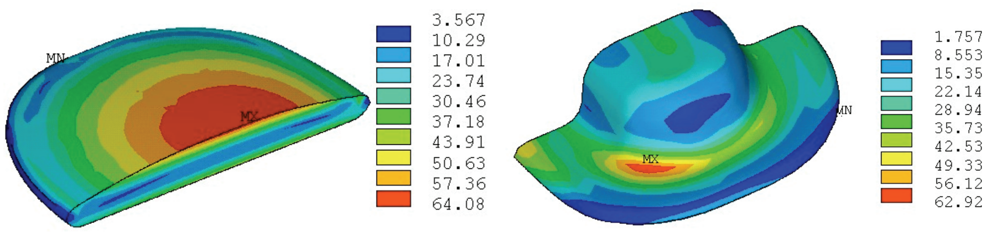

Fig. 3. Von Mises equivalent stress (MPa) in the upper plate at compressive force of 3 kN: integral inlay (left), separate inlay (right)

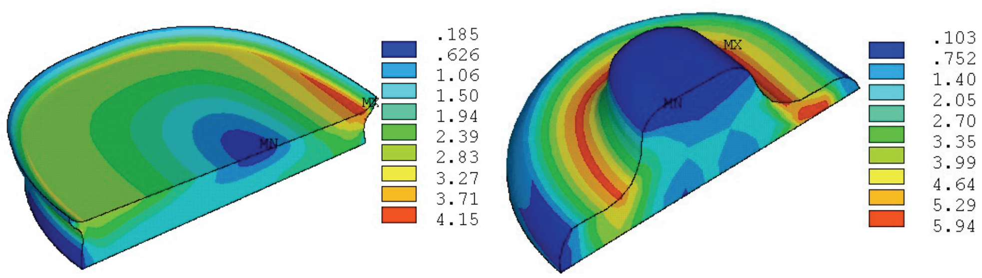

Fig. 4. Von Mises equivalent stress (MPa) in the upper part of the inlay at compressive force of 3 kN: integral inlay (left), separate inlay (right)

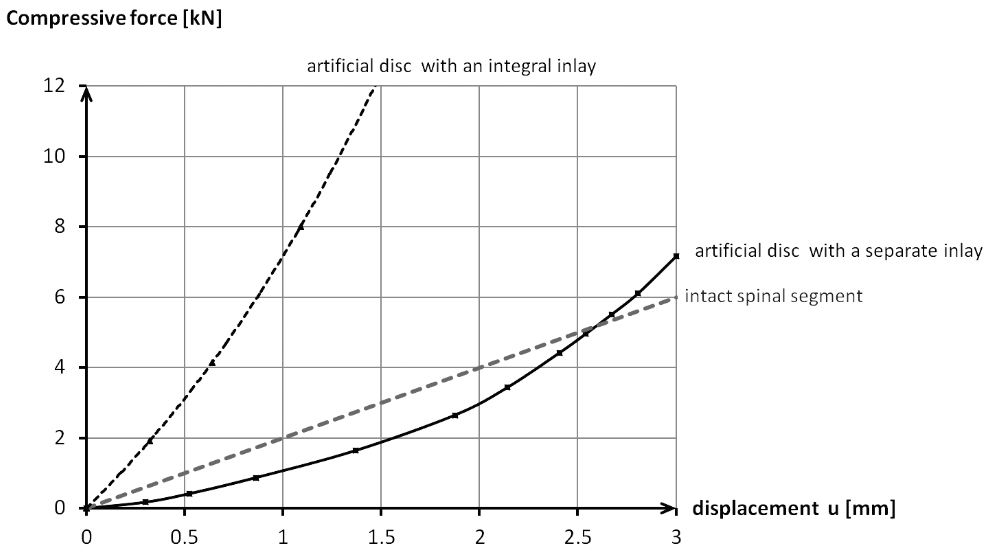

Fig. 5. Curves representing compressive force versus displacement for two types of prostheses

region of the inlay; its magnitude was slightly greater than that for the integral inlay. The maximum values of the equivalent stresses in plates were similar for both types of prostheses (figure 3). The stress levels obtained for the plates were rather low compared to the strength of the Co–Cr–Mo alloy.

3.2. Sagittal bending of the prosthesis with a separate inlay

The deformation of the prosthesis at the maximum value of flexion was shown in figure 6. The inlay protrusions protected the plates against free movements and ensured the prosthesis stability. In the flexion case, the forces of contact between the plates and the inlay were present not only in the anterior part of the inlay, but also at the posterior surfaces of the inlay protrusions. With an increase in the flexion angle, the contact forces at the posterior surfaces of the inlay protrusion decreased, so finally the inclination of the curve shown in figure 6 was smaller for greater angles of flexion (and extension) than for the neutral position of the spinal segment.

The maximum equivalent stress in the inlay for the maximum angle of flexion occurred in the anterior part of the inlay and was slightly greater than that for the maximum angle of extension, where the maximum value (4.7 MPa) occurred in the rear part. These values were comparable to those obtained for the compression force equal to 3 kN (see figure 4).

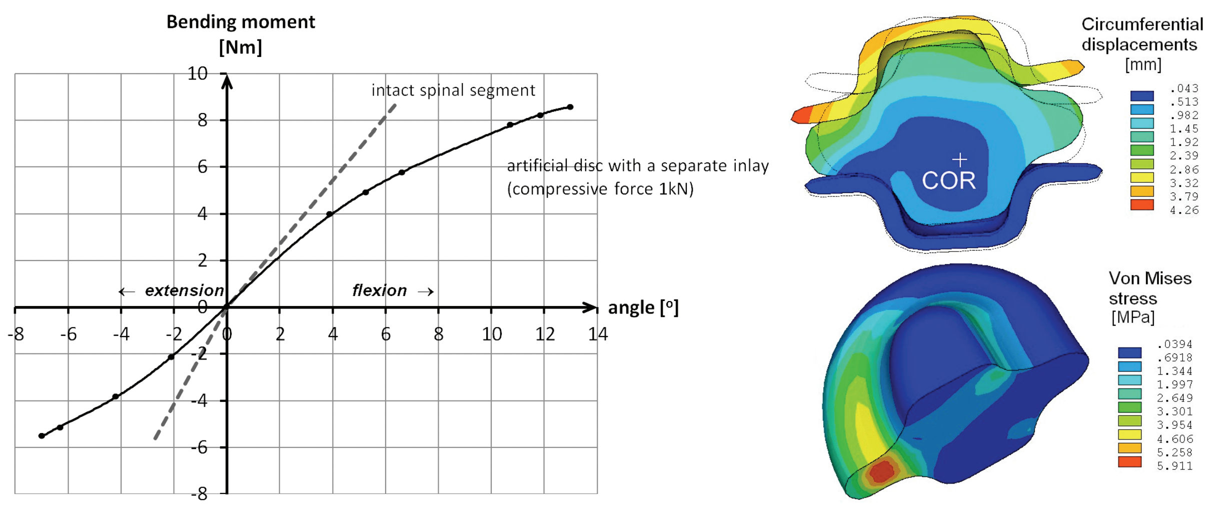

Fig. 6. Curves representing bending moment versus angle of rotation for the prosthesis with a separate inlay under compressive force of 1 kN; the results selected for flexion angle of 13°, COR – centre of rotation

3.3. Sagittal shear of the prosthesis with a separate inlay

The shearing force–horizontal displacement characteristics were almost linear (figure 7). The shear stiffness of the prosthesis with a separate inlay was about 40% smaller than the stiffness of the intact spinal segment [8]. The effort of the inlay was the highest in its central part. The highest equivalent stress value was of the same order as that obtained for the previously analysed cases.

3.4. Axial rotation of the prosthesis with a separate inlay

Rotational stiffness of the artificial disc was too small in comparison with the human intervertebral disc (figure 8). There were two pairs of regions covering the upper and lower surfaces of the inlay which contacted with the plates. The curve representing the torsional moment versus angle of rotation shown in figure 8 is nonlinear and its slope falls for great values of the rotation angles. It is the deformation of protrusions caused

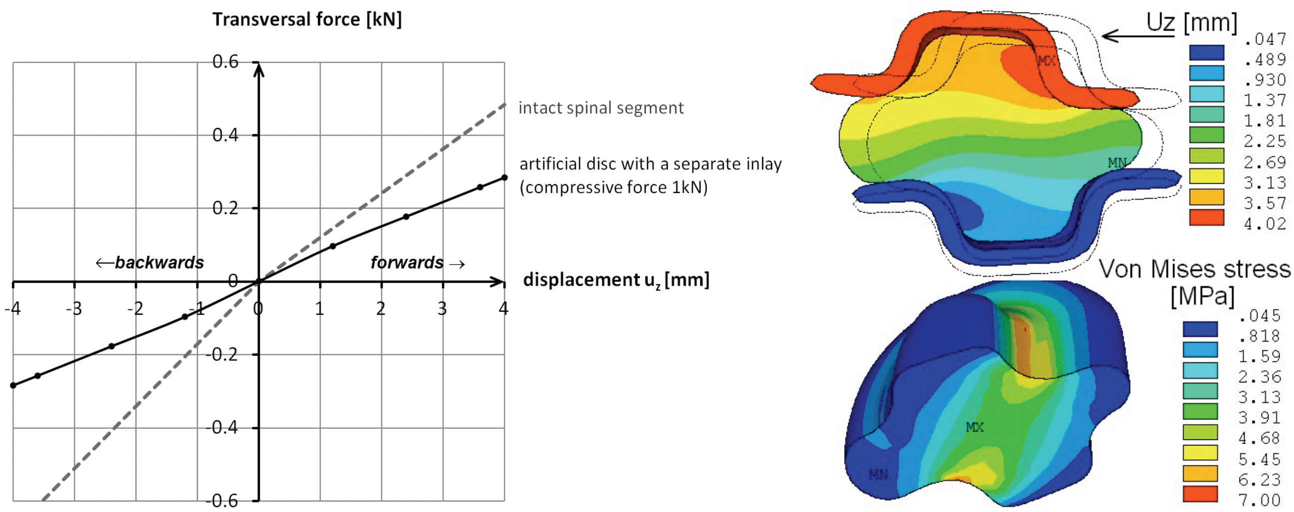

Fig. 7. Transversal force–sagittal displacement u_z characteristics for the prosthesis with a separate inlay under compressive force of 1 kN; selected results for the forward displacement u_z = 4 mm

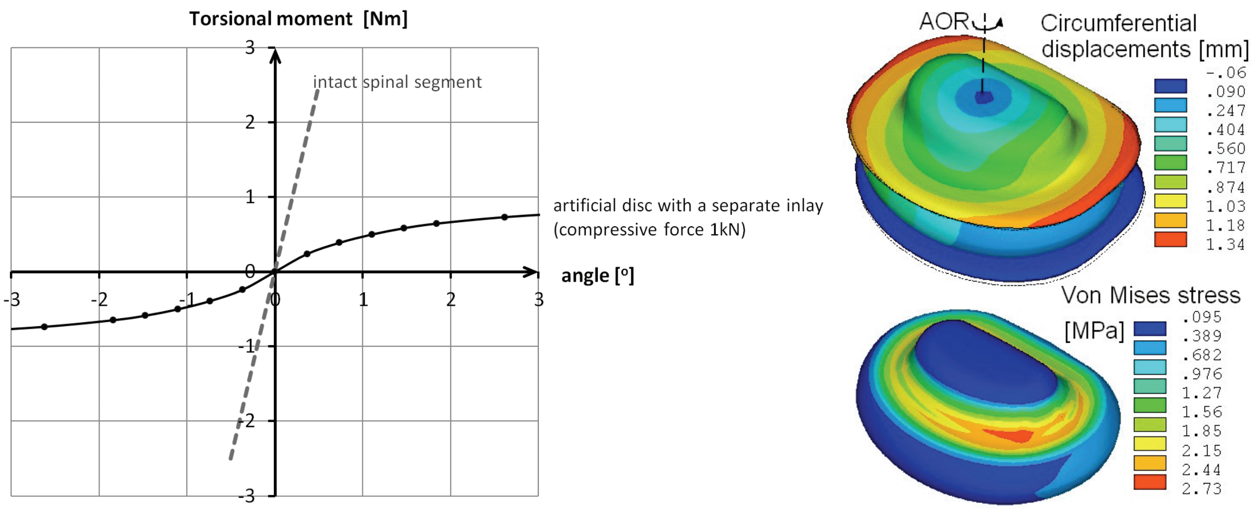

Fig. 8. The curves representing torsional moment versus angle of rotation for the prosthesis with a separate inlay under compressive force of 1 kN; some selected results for 3.6° rotation angle, AOR – axis of rotation

by contact forces that is largely responsible for the slope falling. To simplify, protrusions at the greater values of rotation angles assume cylindrical shape, hence the moment is transferred not by pressure but by friction forces. Contact stress for the assumed range of rotation angle is not high but this occurs because of considerable flexibility of the inlay.

4. Discussion

The analysis presented in this paper shows that it seems to be impossible to imitate exactly the natural disc behaviour for all types of loads and spinal movements by the design with a separate inlay, unless the materials of inlay exhibit isotropic properties.

The greatest difference between the natural and artificial discs occurred in axial rotation. The stiffness of implant was not sufficient but this small stiffness might have exerted positive influence on the facet loads. The ball-joint prosthesis which did not have rotational stiffness was analyzed by SCHMIDT et al. [10]. The inability of the prosthesis to resist torsional moment increased the facet load from 27% to 46% compared to the intact spinal unit. Also the position of the centre of rotation was non-physiological as a consequence of the instability of an artificial device.

The results presented in this paper for other cases proved that the prosthesis with a separate inlay could approximately simulate the behaviour of the human intervertebral disc. For flexion the characteristic was closer to the natural disc compared with the Charite disc [11]. Other numerical analyses [10], [13] indicated that in flexion the ball-joint devices usually increased facet load while no facet forces were observed for the natural disc. Both constrained and unconstrained inlays were taken into account. The authors concluded that constrained artificial discs were more heavily loaded and consequently the load on the facet joints decreased by about 30%. ROHLMANN at al. [12] carried out probabilistic finite element analysis of the L3–L5 segments and concluded that the load on the facet joints depended greatly on individual geometry of vertebrae, the shape and position of implant and the contact parameters. In their simulation, the facet forces in flexion were equal to zero in about 60% cases. To conclude, the device with a separate inlay presented in this paper can resist shear loads and is more stable than the Charite disc and thus may produce smaller load of facet joints in flexion.

The lateral bending is not analyzed, but the results seem to be similar to those obtained for sagittal bending. SCHMIDT et al. [10] noted that for different designs of implants facet loads changed from 4% to even more than 200%. Therefore this case should be subsequently investigated.

The stiffness factors taken from the monograph [8] for the intact spinal segment were only a rough simplification of the spinal segment behaviour. It is nevertheless true that in the case of the artificial disc, the presence of ligaments and facet joints cannot be neglected, especially for bending, shear and axial rotation. The application of too rigid a prosthesis can cause the changes in the kinematics of the spinal segment and may overload the structure of bones and ligaments. On the other hand, too flexible a device can additionally be responsible for compression of the nerve roots. Therefore the influence of the prosthesis on the surrounding structures should be evaluated in the future by using the model of spine with the ligaments, facet joints and the natural discs in adjacent spinal segments.

The results presented relate only to the central position of an artificial disc. The next computations should be supplemented by assuming certain range of the implant position. Different implant positions were analyzed numerically [10], [13]. The results indicated a considerable influence of this parameter on the implant and the surrounding structures' behaviour. At an anterior implant position there was a significant increase in facet load, and the range of extension also increased. At a posterior implant position the facet load decreased and the range of flexion increased.

Finally, the shape of the plates and implant should be optimised according to two criteria. The first one is the possibility of implanting or replacing the inlay in the case of the excessive wear. The second criterion applies to the stability of the inlay position and its capability to resist bending, torsional and shear loads. So, first of all the height of the inlay protrusion and the depth of the plate groove should be considered and also their transverse dimensions and corner radii of the plates and the inlay.

Acknowledgement

The study was financed by the Development Project of the Polish Ministry of Science and Higher Education, No. 15-0028-10.

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