The risk of disc prolapses with complex loading in different degrees of disc degeneration – A finite element analysis

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Abstract

Background. Disc prolapses can result from various complex load situations and degenerative changes in the intervertebral disc. The aim of this finite element study was to find load combinations that would lead to the highest internal stresses in a healthy and in degenerated discs.

Methods. A three-dimensional finite element model of a lumbar spinal segment L4–L5 in different grades of disc degeneration (healthy, mild, moderate, and severe) were generated, in which the disc height reduction, the formation of osteophytes and the increasing of nucleus' compressibility were considered. The intradiscal pressure in the nucleus, the fiber strains, and the shear strains between the annulus and the adjacent endplates under pure and complex loads were investigated.

Results. In all grades of disc degeneration the intradiscal pressure was found to be highest in flexion. The shear and fiber strains predicted a strong increase under lateral bending + flexion for the healthy disc and under axial rotation and lateral bending + axial rotation for all degenerated discs, mostly located in the postero-lateral annulus. Compared to the healthy disc, the mildly degenerated disc indicated an increase of the intradiscal pressure and of the fiber strains, both of 25% in axial rotation. The shear strains showed an increase of 27% in axial rotation + flexion. As from the moderately degenerated disc all measurement parameters strongly decreased.

Interpretation. The results support how specifically changes associated with disc degeneration might contribute to risk of prolapse. Thus, the highest risk of prolapses can be found for healthy and mildly degenerated discs.

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Keywords: Disc prolapse; Disc failure; Disc degeneration; Complex loading; Finite element analysis

1. Introduction

The causes of disc prolapses are multifactorial, including degeneration processes and mechanical effects. Degeneration involves gross structural changes in the intervertebral disc (IVD) and the adjacent endplates (Adams et al., 2002, 2000). One of the first signs of disc degeneration is a reduced hydration capacity (Gunzburg et al., 1992), which can lead to a decrease in intradiscal pressure (IDP) of the nucleus pulposus and in disc height. Other signs of disc degeneration may be the growth of osteophytes and a flattened endplate curvature (Wilke et al., 2006).

Because of these changes, the biomechanical behavior of the disc is inevitably affected. Brown et al. (2002) suggested that the segment stiffness in flexion decreases during the initial stages of degeneration and then increases with the onset of severe disc degeneration. In contrast, Mimura et al. (1994) found that with increasing disc degeneration, the stiffness in flexion/extension increased slightly. Furthermore, they showed that in lateral bending the stiffness rose
strongly, whereas the stiffness was reduced during axial rotation.

Previous in vitro studies demonstrated that a disc may prolapse under certain load combinations of flexion, lateral bending, axial rotation, and axial compression (Edwards et al., 2001; Adams et al., 2000; McNally et al., 1993; Gordon et al., 1991; Adams and Hutton, 1985, 1982b,a). These combinations may produce high pressures in the nucleus and local regions of large stresses in the annulus (McNally and Adams, 1992), with the highest stresses found in the postero-lateral region (Steffen et al., 1998). Several studies have been published investigating the association between the degree of disc degeneration and the susceptibility to disc prolapse (Gallagher, 2002; Adams and Hutton, 1982b). The authors conclude that slightly degenerated discs at lower lumbar levels aged between 30 and 40 years are most susceptible to prolapse. However, the results were not supported by any statistical tests. Only limited data on the relationship between the degree of disc degeneration and the loads required for disc failure have been obtained.

In finite element (FE) analysis, disc degeneration was often simulated by changing the material properties of different disc tissues, such as reducing the incompressibility of the nucleus or increasing of elasticity of the ground substance (Polikeit et al., 2004; Kim et al., 1991). Rohlmann et al. (2006) additionally investigated the changes in geometrical parameters during degeneration. The authors simulated a disc height reduction. Little is known about the effect of other geometrical parameters, such as osteophytes growth or endplate curvature changes. Furthermore, most of the previous FE-studies were only used for simulating pure moments in one of the three anatomical main-planes. However, in the physiological situation a complex state of loading exists. Only few groups investigated the disc behavior under specific load combinations: flexion or extension plus axial rotation (Lu et al., 1996; Shirazi-Adl, 1992, 1991, 1989). The authors found that the maximum fiber strains occurred in the posterior and postero-lateral annulus and are more affected in the inner annulus layer than the outer ones. However, it is not known, which load combination lead to the highest internal pressure and strains in different grades of disc degeneration.

Therefore, the aim of this FE-study was to find load combinations that would lead to the highest internal pressure and strains in differently degenerated IVDs and to determine the locations in which the highest strains occur. We hypothesized that:

(a) the postero-lateral region of the disc would experience the highest internal strains,

(b) the highest internal pressure and strains would occur under complex load situations and

(c) with increasing disc degeneration the internal pressure and strains of the disc and therefore, the risk of getting a disc prolapse would decrease.

2. Methods

2.1. FE-Model

A previously developed non-linear, three-dimensional FE-model of a human lumbar spinal segment L4–L5 was used for this study (Fig. 1). Details of the model development have been given elsewhere (Schmidt et al., 2007a) and are briefly summarized here. The geometry of the FE-model was based on a high resolution computer tomography scan with a slice thickness of 0.75 mm. Spinal structures that could not be recognized using computer tomography were determined by magnetic resonance imaging and histological observations. The modeled vertebrae and the intervening IVD were meshed using eight-node iso-parametric solid elements. The fibers of the annulus fibrosus and the seven spinal ligaments were represented by unidirectional spring elements, which do not offer resistance in compression. The surfaces of the facet joints were simulated by a cartilaginous layer, which was assumed to be multi-linear elastic in compression (Sharma et al., 1995). The contact between the facet joints was simulated by surface to surface contact elements without friction. The orientation and curvature of the facet joints were in accordance to previous observations (Masharawi et al., 2004; Panjabi et al., 1993). The IVD consisted of the nucleus and the surrounding annulus ring. The annulus was assumed to be composite of a homogenous ground substance reinforced by collagen fibers. Eight crisscross fiber layers were defined in radial direction. The fiber angle varied from ±24° to the midcross-sectional plane ventrally to ±46° at the dorsal side according to a previous study (Holzapfel et al., 2005).

Material properties of the different tissues were extracted from the literature and are given elsewhere (Schmidt et al., 2007a). The fluid-like behavior of the nucleus and the hyper-elastic properties of the annulus ground substance were both modeled using the isotropic, incompressible hyper-elastic Mooney–Rivlin $(c_1, c_2)$ formulation.

$$ W = c_1(I_1 - 3) + c_2(I_2 - 3) + \frac{1}{d}(J - 1)^2 $$

(1)

with

$$ c_1, c_2 \quad \text{material constants characterizing the deviatoric deformation of the material} $$

$$ I_1, I_2 \quad \text{first/second invariants of the deviatoric strain tensor} $$

$$ d = 2/K \quad \text{material incompressibility parameter} $$

$$ J = V/V_0 \quad \text{local volume ratio and} $$

$$ K \quad \text{initial bulk modulus of the material}. $$

The material constants for the nucleus and annulus ground substance $c_1$ and $c_2$ were defined using the following approximation for Young’s modulus $E$ (MARC, 2001):

$$ E \cong 6 \cdot (c_1 + c_2) $$

(2)
with
\[ c_2 \approx 0.25 c_1. \]  

(3)

Values of 0.12 and 0.03 for the nucleus and 0.18 and 0.045 for the annulus were used for \( c_1 \) and \( c_2 \), respectively. These values correspond to a Young’s modulus of 0.9 MPa for the nucleus (Smit, 1996) and 1.35 MPa for the annulus (Schmidt et al., 2006).

The stress–strain behavior of the annular fibers were described by a non-linear function, which was obtained from previous reports (Shirazi-Adl et al., 1986). The fiber stiffness increased from inner to outer lamellae (Sharma et al., 1995). All ligaments were described by a non-linear force–deflection curve (Pingel, 1991).

In addition to the healthy disc (grade 0), three different simplified grades of disc degeneration (mild, moderate, and severe) were created (Fig. 2). To model these degenerated discs, the disc height and endplate curvature were decreased, osteophytes formed, the compressibility of the nucleus increased and the fiber and ligament stiffness decreased. This model process was based on a newly developed grading system for IVD degeneration (Wilke et al., 2006). In this grading system the “Height loss” was defined as grade 0: 0%, grade 1: 0–33%, grade 2: 33–66% and grade 3: 66–100% and the length of the osteophytes was defined as grade 0: 0 mm, grade 1: 0–3 mm, grade 2: 3–6 mm and grade 3: >6 mm.

2.2. Changes during disc degeneration

Compared to the healthy disc, we assumed for the mildly, moderately and severely degenerated disc a reduced disc height of 16.5%, 49.5% and 82.5%, respectively, being the mid value for each grade as defined by Wilke et al.

Osteophyte formation

The osteophytes were simulated uniformly around the inferior and superior endplate. The definition of the osteophytes’ length was similar to that for height loss, with the values of 1.5, 4.5 and 7.5 mm being the mid value of grades 1, 2 and 3, respectively (Fig. 2). To our knowledge no material parameters exist describing the behavior of the osteophytes and the soft tissue between the osteophytes. Frozen sections indicated that these materials are similar to the structure of the cancellous bone and the annulus ground substance, respectively.

Facet orientation

The relative orientation of opposing contact surfaces in the facet joints changed due to disc height reduction. In the FE-model of the healthy disc the facet surfaces were oriented parallel to each other (this corresponds to 0°), while in the FE-models with disc degeneration the inferior facet
of L4 was aligned obliquely to the superior facet of L5. The gap between the opposing facet surfaces were widened by mild, moderate and severe disc degeneration creating angles of 0.8°, 1.9° and 2.9°, respectively. These angles were found as a consequence for the reduced disc height.

**Endplate curvature**

In a radiographic study was found that the endplate contour flattens with progressing disc degeneration (Wilke et al., 2005). To modify the endplate in the FE-model in different degrees of disc degeneration, we defined a range of 0% and 100%, whereas 0% correspond the curvature of the healthy disc and 100% corresponded a planar endplate. Wilke and colleagues were not able to quantify the contour changes during disc degeneration. Therefore, we assumed the same percentages as we used for the disc height loss.

**Disc bulge**

In midsagittal frozen sections of degenerated IVDs we found that the disc bulge increased with progressing disc degeneration. To our knowledge, no literature exists quantifying this increase. Therefore, we assumed the same percentages as were used for the disc height loss. The reduced compressibility of the nucleus permitted the inner annulus surface to bulge inward (Adams et al., 2002). Our macroscopical observations indicated no geometrical changes for grade 1, while grade 3 showed nearly the same inner curvature as the outer surface. Therefore, we defined these observations as the limits of our FE-model. For grade 2 a planar interface was assumed.

**Fiber and ligament stiffness**

With decreasing disc height the fibers and most ligaments probably buckle (Rohlmann et al., 2006). In the FE-model a decreased disc height influenced the lengths of the elements representing the collagen fibers and ligaments. Similar to a FE-study of Rohlmann et al. this change in length was compensated by offsetting their non-linear force–deflection curves (Table 1). Hence, when they got their original length, the buckled fibers and ligament became active. In the mildly degenerated disc the anterior and posterior longitudinal ligaments were pre-stressed due to the growing of osteophytes (negative offset value).

**Nucleus’ compressibility**

The Young’s modulus of the nucleus was increased from the material values of the healthy nucleus to the values of the annulus ground substance. From frozen sections it is recognizable that the nucleus and annulus became structurally similar with increasing disc degeneration. Rohlmann et al. (2006) came to the same conclusion in their FE-analysis. The Young’s modulus chosen for the mildly and moderately degenerated nucleus was linear interpolated. The corresponding $c_1$ and $c_2$ coefficients used for Eq. (1) are given in Table 1.

**Annulus ground substance**

It was assumed that disc degeneration has no effect on the material properties of the annulus ground substance. This assumption based on prior in vitro studies (Holzapfel et al., 2005; Ebara et al., 1996). In these studies the authors did not find a significant correlation between the tensile moduli and degeneration related changes.

### Table 1

<table>
<thead>
<tr>
<th>Material properties of the disco-ligamentous tissues in the finite element model in different stages of disc degeneration</th>
</tr>
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<tbody>
<tr>
<td><strong>Grade 0</strong></td>
</tr>
<tr>
<td>Nucleus pulposus</td>
</tr>
<tr>
<td>$c_1 = 0.12$, $c_2 = 0.03$</td>
</tr>
<tr>
<td>Annulus ground substance</td>
</tr>
<tr>
<td>Osteophytes</td>
</tr>
</tbody>
</table>

**Offset values for the non-linear force–deflection curves in mm**

<table>
<thead>
<tr>
<th></th>
<th>$c_1$ (average)</th>
<th>$c_2$ (average)</th>
<th>$c_3$ (average)</th>
<th>$c_4$ (average)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Annulus fibers</td>
<td>0.72</td>
<td>1.68</td>
<td>2.56</td>
<td>3.44</td>
</tr>
<tr>
<td>Lig. long. anterius</td>
<td>–</td>
<td>–</td>
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<tr>
<td>Lig. long. posterius</td>
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<td>Lig. flaval</td>
<td>–</td>
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<td>Ligg. intertransversalia</td>
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<td>Ligg. interspinalia</td>
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<tr>
<td>Capsulae</td>
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<tr>
<td>Ligg. supraspinalia</td>
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<tr>
<td>Ligg. paraspinale</td>
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<tr>
<td>Ligg. supraspinale</td>
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<tr>
<td>Ligg. transversalia</td>
<td>–</td>
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</table>

2.3. Loading and boundary conditions

The inferior endplate of the lower vertebral body was rigidly fixed (Fig. 1). Pure unconstrained moments of 7.5 Nm were applied to an application point above the upper vertebral body. This point was attached to the superior endplate of L4 using rigid beam elements with six degrees of freedom. In that way we were able to transfer moments in the spinal segment. The loading direction was changed in steps of 15° between each pair of the three anatomical main-planes to realize not only pure moments in flexion, extension, lateral bending, and axial rotation.
but also load combinations, for example flexion + lateral bending. The line of action for the resulting moment between two anatomical main planes was an oblique spatial axis. The applied moment about this axis was always 7.5 Nm. All these load cases were additionally combined with an axial compression preload of 500 N. This load was applied as if it was a follower load (Patwardhan et al., 1999). The model was analyzed for large deflection and elastic material properties (ANSYS 11.0; Swanson Analysis, Houston, PA, USA).

2.4. Data analysis

- The range of motion (RoM) and the facet forces were analyzed for validation purposes.
- The IDP in the nucleus was determined as one third of the trace of the stress tensor, i.e. the mean of the three normal stresses. It was assumed that a high nucleus' pressure when combined with small fissures in the annulus can lead to an expulsion of the inner core.
- The tensile strains in the direction of the fibers predicted fiber breakages, which can initiate radial tears (Adams et al., 1985).
- It was found in radiographic studies that the outer annulus separate from the adjacent vertebral bodies and produce peripheral rim lesions (Osti et al., 1990; Hilton et al., 1976). We assumed that these failures are mainly caused by a resulting shear strain. The shear strain $\varepsilon_{RS}$ was calculated using the two strain components $\varepsilon_{xz}$ and $\varepsilon_{yz}$ acting in the transversal plane.

$$\varepsilon_{RS} = \sqrt{\varepsilon_{xz}^2 + \varepsilon_{yz}^2} \quad (4)$$

3. Results

For all grades of degeneration, the model predicted a higher RoM for moments in one of the principal directions, than for moment combinations. The healthy disc model indicated the highest RoM in flexion (6.0°), followed by lateral bending (4.5°), extension (4.3°) and axial rotation (1.8°). Compared to the healthy disc, the mildly degenerated disc indicated an increase of the RoM in axial rotation by 1.0°, in flexion by 0.5° and in extension by 0.1°. However, lateral bending generated a slight decrease (0.7°). In the moderately degenerated disc, only axial rotation predicted a larger RoM (0.5°) compared to the healthy disc. For flexion, extension, and lateral bending a strong decrease was determined (mean: 2.1°). For the severely degenerated disc the RoM strongly decreased in all load directions.

Similar to the RoM, the model predicted higher facet forces for moments in one of the three anatomical main-planes than for moment combinations. The maximum contact forces were determined under axial rotation (105 N), followed by extension (50 N), and lateral bending (36 N). The facet joints remained unloaded during flexion. For increasing disc degeneration the model predicted higher forces for axial rotation whereas for extension and lateral bending we found increased facet forces for mildly degenerated discs, which decreased with further degeneration.

The IDP for the healthy disc model was found to be highest in flexion (0.71 MPa), followed by extension (0.62 MPa), axial rotation (0.57 MPa) and lateral bending (0.44 MPa) (Fig. 3). Except axial rotation, the mildly degenerated disc indicated in all load directions only slight differences in IDP compared to the healthy disc. Axial rotation resulted in an IDP increase of 25%. As from the moderately degenerated disc the IDP strongly decreased in all load directions. The model predicted a higher IDP for moments in one of the three anatomical main-planes than for moment combinations.

The healthy disc model predicted largest fiber strains (19.8%) under a combination of lateral bending + left axial rotation (Fig. 4). Compared to the healthy disc, the mildly degenerated disc model indicated an increase of the fiber strains from 11.9% to 16.1% for axial rotation. Combined moments did not lead to an increase of fiber strains compared to moments in one of the three anatomical main-planes. In the moderately degenerated disc the fiber strains strongly decreased in all load directions compared to the healthy disc. Only loads combined with an axial rotation caused fiber strains. The severely degenerated disc model predicted unloaded fibers in all load directions. The maximum fiber strains for the healthy disc model were determined in the postero-lateral annulus region (Fig. 5). During disc degeneration the region of maximum fiber strains increased.

For the healthy disc model, the maximum shear strain was determined under load combinations of lateral bending + flexion (44.7%) and lateral bending + extension (45.5%) (Fig. 6). For the mildly and moderately degenerated disc models, the maximum shear strains were determined under the load combination of axial rotation + flexion (grade 1: 47%, grade 2: 39%). In the severely degenerated disc, only the combination of axial rotation + flexion resulted in an increase of the shear strains (up to 29%). The maximum shear strains were mostly located in the posterior and postero-lateral annulus region (Fig. 7). With increasing disc degeneration the maximum strain transferred to the postero-lateral annulus.

4. Discussion

Under the assumptions of this simplified FE-model the results yielded some general suggestions, which might be important in clarifying the cause of disc prolapses in a healthy disc compared to disc with degeneration.

The results from our FE-model supported the hypothesis (a) in that the highest internal strains occurred in the
The data are consistent with previous clinical and in vitro findings (Ebeling and Reulen, 1992; Spangfort, 1972). Hypothesis (b) was supported by the results in that the highest internal strains were located under combined imposed loads for the fiber and shear strains in the annulus. However, for the IDP in the nucleus hypothesis (b) was not supported by the data. IDP was not strongly influenced by the load combinations. Rather, the IDP was dependent on the total RoM. Also partially supported by the data was the hypothesis (c) in that increasing disc degeneration was related to a decreased risk of disc prolapse. The model predicted that the risk of disc prolapse initially increases for the mildly degenerated disc and subsequently decreases for the moderately and severely degenerated disc, which is consistent with previous in vitro studies (Gallagher, 2002; Adams and Hutton, 1982b).

Especially for axial rotation it was found that the mildly degenerated disc generated an increase of the IDP.

Fig. 3. Intradiscal pressure ($p_{ID}$) in the nucleus pulposus in dependent of disc degeneration under pure and combined moments of 7.5 N m and an axial compression preload of 500 N. The pressure is diagrammed in cylindrical coordinates – IDP is shown radially and the applied load in the circumference. Left: lateral bending + flexion and extension, middle: axial rotation + flexion and extension, right: right lateral bending + right and left axial rotation. (For interpretation of the references to colour this figure legend, the reader is referred to the web version of this article).

Fig. 4. Maximum tensile strain of the annulus fibers in dependent of disc degeneration under pure and combined moments of 7.5 N m and axial compression preload of 500 N. The strain is diagrammed in cylindrical coordinates – strain is shown radially and the applied load in the circumference. Left: lateral bending + flexion and extension, middle: axial rotation + flexion and extension, right: lateral bending + right and left axial rotation. Except for axial rotation, for all load cases the moderately degenerated disc and for all load cases the severely degenerated disc led to unloaded fibers. (For interpretation of the references to colour this figure legend, the reader is referred to the web version of this article).
pared to the healthy disc. An axial preload alone would resulted in an IDP decrease from 0.34 (grade 0) to 0.29 MPa (grade 1). Therefore, the indicated increase of the IDP was caused by the pure rotation moment when moments and axial preload are simultaneously applied. This could be due to the RoM increase between grade 0 and 1.

A degenerated nucleus is characterized by dehydroge- nated tissue, and by an increase of compressibility. In addition, the slack of the ligaments increases and the pre-stress in the flaval ligaments decreases. These characteristics may also lead to a reduced IDP in the moderately and severely degenerated discs, which correlates with previous in vitro studies (Adams et al., 2002; McNally and Adams, 1992).

In experimental studies it was found that the ultimate tensile strain of fibers is 10–25% (Holzapfel et al., 2005; Haut, 1983; Betsch and Baer, 1980). This suggests that under lateral bending plus left axial rotation, fibers in the postero-lateral region may be susceptible to rupture. However, the fiber strains would not substantially change when higher moments are applied. With increasing moment the fiber strains and the facet forces are increasing. At some certain point the contact of the facet joints is completely closed. Therefore, the range of motion is only slightly increasing and the maximum fiber strain reaches a limit. Because of the closed contact the facet forces are increasing rapidly under the strain plateau.

The fibers strains are higher under axial rotation for the mildly degenerated disc when compared to the healthy disc. This might be due to the fiber angle, which decreases in relation to the transversal plane during disc height reduc- tion. Thus they became more effective in axial rotation than

Fig. 5. Locations of predicted maximum fiber strains in the annulus fibrosus under combined moments in dependent of disc degeneration. Regions of the annulus, which were larger than 90% of this peak strain, were depicted as colored areas. Except for axial rotation, for all load cases the moderately degenerated disc and for all load cases the severely degenerated disc led to unloaded fibers. (For interpretation of the references to colour this figure legend, the reader is referred to the web version of this article).
in bending. It should be noted that the disc is modeled with the same number of elements in disc height in each degree of disc degeneration. Therefore, the angle of the fibers decreased with increasing disc degeneration. As from the moderately degenerated disc, only load combinations with an axial rotation result in loaded fibers. The resulting strains, however, were much less than those of the healthy disc. Therefore, we believe, that these small strains barely destroy the fiber construct.

Very high shear strains were found between the annulus and the adjacent endplates. These values are likely to induce failure at this level. Clinical studies showed that disc prolapses mostly occurred in the postero-lateral region of the disc (Ebeling and Reulen, 1992; Spangfort, 1972) mostly at the adjacent inferior endplates (Dillon et al., 1983). This type of fissures might be due to high shear strains between annulus and adjacent endplates. The healthy and all degenerated disc models predicted the same trends.

The healthy disc model was validated in previous studies on RoM, IDP and disc bulge (Schmidt et al., 2007a,b). The FE-models describing the different degrees of disc degeneration were compared to the RoM determined in previous in vitro studies (Brown et al., 2002; Mimura et al., 1994). Mimura et al. found that the RoM is reduced for degenerated discs in flexion/extension and lateral bending, while it is increased for axial rotation. Our FE-model predicts the same trends for lateral bending and axial rotation. In flexion and extension, however, our results differed from these values measured by Mimura et al. While we calculated an increase in RoM, Mimura et al. determined a reduced flexibility. However, Brown et al. measured a slight increase in RoM, which is in good comparison to our findings. For the healthy disc model our calculated facet forces are in good agreement with prior in vitro findings (Wilson et al., 2006). However, no in vitro data exits describing the facet forces in dependant of disc degeneration. For that reason we compared our results with prior numerical findings (Rohlmann et al., 2006). In this study the authors find the same trends as our results predicted.

4.1. Assumptions and limitations

The described methodology has quite a few limitations and uncertainties regarding the simulation of different grades of disc degeneration. The problem is that the effects of degeneration on the mechanical properties and geometrical changes of spinal ligamentous tissues are still unknown. Therefore, several assumptions were necessary. We derived these assumptions phenomenological from extensively macroscopical and radiographic studies comparing the healthy and degenerated discs. All these assumption have an influence on the results. Hence, the results of the current study can only indicate trends and do not necessarily represent the correct values. However, we intended to perform a first step into the simulations of disc degeneration.

All degeneration characteristics were carried out to the same magnitude in each of the degeneration grades. However, the sign of degeneration may individually differ (Wilke et al., 2006). Furthermore, degenerative changes of the facets joints remained unconsidered. The effect of tears in the annulus also seen during degeneration (Fujita et al., 1997; Osti et al., 1992) was not evaluated currently.

Still, there is no in vitro study reported to date, which examined the material parameters of the nucleus, osteophytes, and the soft tissue between the osteophytes during disc degeneration. In a first approximation we assumed a
linear interpolation between the material parameters of grades 0 and 3. Seemingly, disc degeneration is not linear. However, a non-linear interpolation would require additional other assumptions, which will be even harder to proof by in vitro studies. Own sensibility studies showed that the influence of the material properties of the osteophytes on the strain behavior inside the disc is not essential, whereas the material of the soft tissue between the osteophytes affected the strains in the disc strongly. Therefore, results of this study should be interpreted with care.

This study has not looked at the position of the nucleus within the disc. This position can change depending on the degeneration of the disc and may influence the IDP and the overall results of the study. Furthermore, fluid movement and fluid loss are particularly important in the risk of disc prolapse. This has not been accounted for in this study.

The investigated loads are small compared to those anticipated in lifting activities, which are known as a prevalent cause of disc failure. However, with the applied moments of 7.5 Nm we were able to compare our results with published in vitro data of Mimura et al. Furthermore, shear forces, which have also been implicated in disc failure and degeneration, were neglected. Up to date there are no proper studies dealing with the experimental evaluation of shear forces in the IVD. Therefore, we would no have been able to validate the model correctly. However, this lack of completeness is object to future studies.

5. Conclusions

Our simplified FE-model predicts that the highest internal pressure and strains in the disc can be found for healthy
and mildly degenerated discs under complex loading conditions, such as combinations of lateral bending + axial rotation and lateral bending + flexion for the healthy disc and under pure axial rotation and a combination of lateral bending + axial rotation for the mildly degenerated disc. The maximum strains are located in the posterior and postero-lateral annulus region, which corresponds to clinical observations. Furthermore, it is predicted that the mildly degenerated disc has a higher risk for disc prolapses than strongly degenerated discs.

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