Tobias Pitzen
Fred H. Geisler
Dieter Matthis
Hans Müller-Storz
Wolf-Ingo Steudel

Motion of threaded cages in posterior lumbar interbody fusion

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T. Pitzen (✉) · W.-I. Steudel
Department of Neurosurgery,
Saarland University Hospital,
66421 Homburg, Germany
e-mail: ncwste@med-rz.uni-sb.de,
Tel.: +49-6841-164400,
Fax: +49-6841-164480

F. H. Geisler
Chicago Institute of Neurosurgery
and Neuroresearch,
Chicago, Illinois, USA

D. Matthis · H. Müller-Storz
Steinbeis-Transferzentrum,
Schwingungs- und Biomechanik,
Offenburg, Germany

Abstract A high rate of pseudarthrosis and a high overall rate of implant migration requiring surgical revision has been reported following posterior lumbar interbody fusion using BAK threaded cages. The high rate of both pseudarthrosis and implant migration may be due to poor fixation of the implant. The purpose of this study was to analyse the motion of threaded cages in posterior lumbar interbody fusion. Six cadaveric human lumbar spine segments (three L2/3 and three L4/5 segments) were prepared for biomechanical testing. The segments were tested, without preload, under forces of axial compression (600 N), torsion (25 Nm) and shearing force (250 N). The tests were performed first with the segments in an intact state, and subsequently following instrumented stabilisation with two BAK cages via

a posterior approach. These results were compared with those of a finite element model simulating the effects of identical forces on the segments with constructs. As the results were comparable, the finite element model was used for analysing the motion of BAK cages within the disc space. Motion of the implants was not seen in compression. In torsion, a rolling motion was noted, with a range of motion of 10.6° around the central axis of the implant when left/right torsion (25 Nm) was applied. The way the implants move within the segment may be due to their special shape: the thread of the implants can not prevent the BAK cages rolling within the disc space.

Key words Spinal fusion · Implants · Biomechanics · Computer simulation

Introduction

Posterior lumbar interbody fusion (PLIF) was first described by Briggs and Milligan in 1944 [4] and modified by Cloward in 1953 [6]. PLIF is a surgically sophisticated technique for arthrodesis and stabilisation within the intradiscal space of an unstable lumbar segment. It consists of a generous posterior decompression of the neural structures and then a discectomy and anterior interbody arthrodesis. The intradiscal work is performed between the exiting nerve root and the retracted transversal nerve root and thecal sac at the lumbar motion segment to be stabilised. Indications for this procedure include degener-

ative lumbar instability and spondylolisthesis [6, 11, 12, 15]. This surgical approach has the advantage over posterior arthrodesis of obtaining a mature fusion in the anterior column, which bears 80% of the axial load and is under compression in an upright posture [20]. Cloward's technique [6] consisted of interbody fusion using autologous iliac crest bone grafts without additional internal stabilisation. The major disadvantage of this technique is the minimal initial internal stabilisation. Furthermore, the stiffness of these constructs decreases in the first few months during the graft absorption and initial integration phase. Graft-related complications with this technique have been reported to range from 3 to 18% [5, 16, 22]. To improve these initial results with PLIF operations, a vari-

ety of implants aimed at providing increased initial stability of the motion segment have been developed to reduce graft-related complications and improve clinical outcome [2, 3, 5, 11, 12, 13, 15].

The surgeon currently has a wide variety of internal stabilisation options potentially available [2, 3, 5, 11, 12, 13, 15]. The initial stability or stiffness of the construct depends on ligamentous stability and bone mineral density. The implants utilised should also serve as stabilising structures.

The use of threaded cages for PLIF usually calls for laminectomy, discectomy and removal of the endplates for application of the implants within the disc space. Destruction of these important elements, as well as of the anterior and posterior part of the lumbar spine, can be expected to increase the range of motion of a spinal segment, especially in torsion and shearing. Moreover, it should be expected that the cylindrical shape of the threaded implants may not be able to stabilise the segment, especially against torsion. The combination of these biomechanical properties of the surgically treated spine and the implant itself could culminate in a high rate of implant migration and pseudarthrosis. In fact, the reported rate of implant migration is 2.7%, with 1.2% requiring reoperation [15], and the reported rate of pseudarthrosis is 32% in single-segment cases and 79% in bisegmental cases [13].

Thus, the aim of this study was to analyse the movements of threaded implants following PLIF in a human lumbar spine segment using a combined biomechanical cadaver and finite element approach.

Materials and methods

Biomechanical study

The biomechanical part of the study was conducted to validate the finite element model. Six cadaveric lumbar spine segments (three segments L2/3 and three segments L4/5) were explanted from six fresh human cadavers, frozen at -20°C and thawed before preparation and biomechanical testing. The mean age of the patients at the time of death was 64.8 years, ranging from 39 to 86 years. The posterior lumbar muscles were removed carefully, without damaging the ligamentous structures or joints.

Self-tapping screws were inserted into the endplates of the vertebrae on the superior and inferior ends of the prepared motion segment. The two ends, along with the screws, were then potted in polymethylmethacrylate (PMMA, Technovit 3040, Heraeus-Kulzer, Wehrheim, Germany) for rigid fixation of the motion segment. Biomechanical testing was then performed in a tension/torsion-testing machine (MMT 0005, Schenk, Germany). The PMMA was rigidly fixed in the testing apparatus, and the specimens were kept moist during testing [1].

For mechanical testing, the specimens were positioned horizontally for axial compression and torsion, and were positioned at an angle of 45° between the inferior endplate of the inferior body and the horizontal plane for shear forces. By positioning the specimen in this manner, the shear force was applied to the disc space as a combined compression-shear force, as pure shearing forces in the human body at the lumbar level do not occur under physiolog-

Fig. 1 The specimens underwent posterior decompression (removal of the upper lamina and the medial facet of the facet joints) and discectomy. A bone graft was placed in the anterior third of the former disc space and the intradiscal hardware was inserted in distraction

ical conditions [1]. The axial compression used 600 N, torsion (left and right) used 25 Nm and shear force used 250 N without preload [1, 7, 11].

After measurements were made on the intact motion segment, the lamina of the upper vertebra and the medial facet of the facet joints were removed. Next, a discectomy was performed, leaving the lateral parts of the anulus intact. An autologous bone graft, made from chips of the resected lamina, was placed in the anterior third of the disc space. The two BAK threaded cages were inserted into the disc space with prior distraction of the disc space using the commercially available instruments and according to the manufacturer's recommended sizing and surgical techniques (Fig. 1).

Each portion of biomechanical testing was carried out three times, with the third cycle used for measurement and statistical analysis. Mean stiffness and standard deviation for all modalities were calculated.

Finite element analysis

ANSYS 5.4, a finite element programme (Swanson Analysis Systems, Inc., Houston, Tex., USA) was used to create the mathematical model of the lumbar spinal motion segment with and without internal stabilisation.

Two separate finite element models were used: one for the intact segment and the other for the segment after posterior decompression and internal stabilisation with BAK threaded cages (Fig. 2, Fig. 3). These models are mathematical representations of the initial lumbar segment and its immediate postoperative stability for the internal stabilisation used.

The shape of a lumbar segment was reconstructed from data obtained after computed tomography scans of a human L3/4 segment as detailed anatomy, and interpreted as anisotropic finite element model, based on the work done by Goel et al. [9, 10] and Shirazi-Adl et al. [17]. The thickness of the endplates and cortical shell were set to 0.25 and 0.4 mm respectively [18]. The structure of the anulus fibrosus was simulated by a ground substance embedded with 768 fibres at alternating orientations of 30° [10]. The nucleus pulposus was specified as an incompressible structure [10]. The two facet joints were simulated as gap elements, with all essential ligaments added to the bony structures, using material properties from the literature [9, 10, 14, 17, 19]. The finite element model of the intact L3/4 motion segment consisted of 7433 single elements.

Surgical decompression, arthrodesis bone and internal stabilisation hardware were added to the intact segment model to simulate the PLIF procedure. First, the posterior elements were removed from the intact model to simulate the posterior decompression required to perform the implantation of the intradiscal hard-

Fig. 2 Finite element model of two BAK cages

ware. Following discectomy, autologous cancellous bone graft was placed in the anterior third of the disc space in PLIF procedures. The cancellous autologous bone graft placed between the anterior

third of the vertebral bodies is modeled as not connected to the endplates, as it has not healed into a fusion mass at the time the operation is concluded. Two (13×20 -mm) BAK cages were placed between the vertebral bodies with the posterior ends 1 mm from the posterior wall of the vertebral bodies, and 1.5 mm separating the cages. The thread of the cages perforated the endplates of the vertebral bodies by 2.3 mm, but did not reach the lateral portion of the anulus.

The BAK threaded cage model simulated the distraction with pre-tension of 5% onto the anular fibres. The bone-implant interface was modeled by point-to-surface contact elements. These elements are able to transmit compression forces but not tension. The PLIF procedures mimic the clinical situation, with the posterior part of the anulus and the nucleus pulposus removed for insertion of the interbody implants, but the lateral and anterior parts of the anulus left intact. The interbody implants are not attached to the anulus. The finite element model L3/4 with BAK threaded cage stabilisation consisted of 6518 single elements (Fig. 3).

The finite element model was loaded using the same forces and moments that were used in the biomechanical part of the study, and range of motion and stiffness of the segment were calculated.

The predictions of the finite element study were compared to the results from the biomechanical part of the study. As the results were comparable, the finite element model was used to simulate and predict the motion of the implants in different loading cases.

Motion of the implants was measured by comparing the position of the implants when no forces or moments were applied with the position they had when maximal forces or moments were applied.

Fig. 3 Finite element model of L3/4 with two BAK cages inserted via a posterior approach

Table 1 Biomechanics of a lumbar motion segment in an intact state and following posterior lumbar interbody fusion stabilised with two BAK threaded cages: comparison of the results of a cadaver study with those of a finite element model

	Cadaver study	Finite element analysis
Intact specimen		
Compression (N/mm)	772 $\pm$ 121	699
Torsion (Nm/°)	3.74 $\pm$ 1.11	7.7
Shear (N/mm)	102 $\pm$ 58	173
After stabilisation		
Compression (N/mm)	549 $\pm$ 357	678
Torsion (Nm/°)	1.97 $\pm$ 0.86	2.5
Shear (N/mm)	100 $\pm$ 48	143

Results

Cadaver biomechanical study

The mean stiffness of the intact specimens was 772 ($\pm$ 121) N/mm for compression, 3.74 ($\pm$ 1.11) Nm/° for torsion and 102 ($\pm$ 58) N/mm for shear forces. There were no noteworthy differences between the group of L2/3 versus L4/5 lumbar segments.

Following PLIF with internal stabilisation with BAK threaded cages, the stiffness was 549 ($\pm$ 357) N/mm for axial compression, 1.97 ($\pm$ 0.86) Nm/° for torsion and 100 ($\pm$ 48) N/mm for shearing forces (Table 1).

A two tailed *t*-test was done to compare the results of the intact to that of the altered specimens. There was no significant difference between the intact and altered spec-

imens in either compression ($P = 0.14$), shearing ($P = 0.91$) or torsion ($P = 0.06$).

Finite element analysis

The finite element model predicted that the intact motion segment would have a stiffness of 699 N/mm in compression, 7.7 Nm/° in torsion and 173 N/mm in shear. The model predicted that the BAK threaded cage stabilisation would have a stiffness of 678 N/mm in compression, 2.5 Nm/° in torsion and 143 N/mm in shear (Table 1).

Note the similarity of the cadaver biomechanical measurements and the predictions made from the finite element model. Except for the results of the intact specimens under torsion, the results of the finite element method are always within one standard deviation of the results of the biomechanical analysis.

Thus, we were encouraged by the general agreement between the cadaveric measurements and the finite element analysis to use this finite element model for prediction of motion of the threaded implants in the disc space.

Motion of the implants

Motion of the implants was not seen in compression. Motion was most pronounced in torsion. It can be described as a rolling movement, with a range of motion of 10.6° around the central axis of the implant when left/ right torsion (25 Nm) is applied (Fig. 4).

In shearing, there was another direction of movement due to the lack of posterior elements to stabilise against shearing. Tilting of 0.3 mm was seen at the posterior edge of the implants.

Fig. 4 Motility of the BAK cages in torsion: **A** 0 Nm, **B** 25 Nm

Discussion

The finite element method is currently gaining greater acceptance in spinal research. However, validation by biomechanical analysis is necessary to ensure accuracy of the predictions from a finite element model [9]. The finite element model presented in this study was validated by biomechanical analysis performed using identical forces and moments for spinal loading. Obviously, the results from the biomechanical and the finite element analysis are not identical, but they are very similar, and therefore comparable. The results of the surgically altered finite element model for each loading case are within one standard deviation of the results of the biomechanical cadaver analysis, and therefore can be trusted. This method of validating a finite element analysis has been described previously [8]. The minor differences – especially in torsion – can be easily explained: in general the differences are due to biological variability in the sample and the fact that the material properties are only approximations for the sample analysed. Moreover, the disc model used in the finite element model represents a healthy disc without any signs of degeneration, in contrast to the discs of the cadaver segments used in the biomechanical part of the analysis. Thus, the model presented here can be used to answer different questions in spinal research such as, for example, motion of spinal implants within a surgically treated segment.

Recommendations for testing of spinal implants are still under discussion [1, 21] and strongly depend on which aspect of the issue is addressed. The present study used pure moments and forces with no preload and without taking account of muscle forces; it therefore did not represent the real physiological loading of the lumbar spine. Application of preload as well as simulation of muscle forces are important parameters for spinal stabilisation *in vivo* [21]; however, as the physiological preload and muscle forces are not known, they had to be omitted from the study. Moreover, the use of pure bending moments for biomechanical testing of spinal implants was not possible with our testing machine. A combination of compression, compression-shearing and torsion was therefore used. Testing in compression seems to be important, as threaded cages are fixed into the cancellous bone and therefore settling may occur. As pure shearing forces do not exist in physiologic conditions [1], testing in combined compression-shearing may mimic *in vivo* conditions. Finally, biomechanical testing of the implants was not the goal of the study; the procedure was carried out to validate the finite element study.

Migration of threaded implants within a treated segment has been observed by Kuslich et al. in 1.2% of cases requiring surgical revision and in 1.5% of cases without the need for revision [15]. Moreover, significant motion of interbody implants may result in delayed fusion or even pseudarthrosis, requiring repeated surgery. Thus, migration of implants within a surgically treated segment is worth examination.

The way the implants move within the segment may be due to their special shape: the cylindrical, threaded shape of the implants is conducive to a rolling motion within the disc space in torsion. Tilting of the implants in shearing, however, may be due primarily to the lack of posterior elements in the model used and, secondarily to the fact that shearing forces were not applied parallel to the disc space. To the authors' knowledge, pure shearing forces do not exist *in vivo* in the human lumbar spine. Thus, we applied shearing forces at an angle of 45° to the motion segment, representing a combination of shearing and compression. Application of shearing forces in the way described onto the segment compresses the anterior edge of the implant and the adjacent vertebral bodies, and therefore results in an opening of the posterior parts of the disc space. This is even more pronounced by the fact that the posterior parts of the spinal column have been destroyed.

Movement of threaded implants within a segment is most pronounced in the face of torsion and shearing forces. However, the results of the current study exclusively apply to PLIF with the need for laminectomy, arthrectomy and removal of the inner parts of the disc. Therefore, the results can not be used to predict movement of threaded implants following anterior lumbar interbody fusion (ALIF). In ALIF procedures, the posterior parts of the spinal column remain intact, stabilising the spine against torsion and shearing. It may be expected that movement of the implants used would be diminished if they are used for ALIF. Similarly, if additional posterior osteosynthesis is used in PLIF – a procedure to reconstruct the posterior elements of the spine – additional stabilisation is provided, and, again, it may be expected that movement of the implants will be reduced.

Despite the limitations of the model used for this study, the results may explain the results of clinical studies on the use of threaded implants in spinal surgery. There are two clinical aspects that need to be discussed with regard to the results of the present study: the rate of pseudarthrosis and the rate of implant migration following PLIF with threaded implants. The rate of pseudarthrosis is described to be as high as 79% in bisegmental procedures [13]. Migration of threaded implants within a fused segment has been observed in 1.2% of cases requiring surgical revision and in 1.5% of cases without the need for revision [15]. Both the high rate of pseudarthrosis and the rate of implant migration may be due to movement of threaded implants preventing solid bony fusion.

Conclusion

Motion of threaded implants used for posterior lumbar interbody fusion is most pronounced in torsion. The thread of the implants can not prevent the implants rolling within the disc space.

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