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The influence of cancellous bone density on load sharing in human lumbar spine: a comparison between an intact and a surgically altered motion segment

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Abstract The aim of the current study is twofold: first, to compare load sharing in compression between an intact and a surgically repaired lumbar spine motion segment L3/4 using a biomechanically validated finite element approach; second, to analyse the influence of bone mineral density on load sharing. Six cadaveric human lumbar spine segments (three segments L2/3 and three segments L4/5) were taken from fresh human cadavers. The intact segments were tested under axial compression of 600 N, first without preload and then following instrumented stabilisation. These results were compared to a finite element model simulating the effect of identical force on the intact segments and the segments with constructs. The predictions of both the intact and the surgically altered finite element model were always within one standard deviation of the mean stiffness as analysed by the biomechanical study. Thus, the finite element model was used to analyse load sharing under compression in an intact and a surgically repaired human lumbar spine segment model, using a variety of E moduli for cancellous bone of the vertebral bodies. In both the intact and the surgically altered model, 89% of the applied load passed through the vertebral bodies and the disc if an E modulus of 25 MPa was used for cancellous bone density. Using 10 MPa – representing soft, osteoporotic bone – this percentage decreased, but it increased using 100 MPa in both the intact and the altered segment. Thus, it is concluded that reconstruction of both the disc and the posterior elements with the implants used in the study recreates the ability of the spine to act as a load-sharing construction in compression. The similarity in load sharing between normal and instrumented spines appears to depend on assumed bone density, and it may also depend on applied load and loading history.

Keywords Lumbar spine · Segmental fixation · Bone · Biomechanics · Finite element analysis

Introduction

Little is known about the exact distribution of forces within healthy lumbar spine due to its complex geometry and physiology. Furthermore, defects by trauma or tumorous destruction may influence spinal biomechanics to a great extent. Surgery also changes the biomechanics of the spine, and the true way in which the spine behaves following, for instance, decompression and stabilisation can only be estimated.

Loads applied onto the spine are shared among spinal components: this is called spinal load sharing. We know that lumbar spinal load sharing takes place, passing 80–96% of the applied load through the anterior part of the spine, and the remainder through the posterior elements [2, 7, 8, 10, 12, 16]. The mechanical importance of the posterior structures of the spine has recently been evaluated by finite element models. For example, the pedicle functions as a structural buttress, providing support to the posterior wall of the vertebral body [27]. Goel [8] predicted load sharing in the cervical spine, such that 88% of compression forces pass the disc and the adjacent bodies, the remainder passing through the posterior elements.

Thus, both the anterior and posterior part of the human spine have an important role for spinal integrity. However, even if the anterior part seems to play the major role in spinal load sharing, clinical experience has shown that the function of the posterior elements can not be neglected. Destruction of the posterior parts may result in kyphotic deformity in cases of spinal trauma or following laminectomy [25, 26]. In children, posterior decompression may lead to severe scoliosis of the thoracic and lumbar spine [5]. Thus, reconstruction of the posterior spinal elements should be performed, following extensive decompression. For surgical reconstruction of the destabilised spine, different implants have been developed.

Implants used for spinal reconstruction should mimic the physiological load-sharing properties of an intact spinal segment, at least partially, for the biomechanical and clinical reasons mentioned above. With special reference to load sharing in an intact spinal segment, the surgically altered segment should be close to this. Goel and coworkers have shown, using a finite element model of the human lumbar spine, that 96% of an applied compression force is passed through the vertebral bodies and the disc. The facet joints are responsible for taking 4% [10]. When an interbody bone graft and posterior pedicle screw-plate stabilisation was simulated in this model, the bone graft was found to be responsible for 80% of the load bearing, and the screw-plate stabilisation for 20% [10]. This stabilisation procedure mimics the character of the spine as a load-sharing construction. If, however, a posterior screw-plate osteosynthesis is used without interbody bone graft, the load through the plate increases to 40% of the applied compression force. Furthermore, if the stiffness of the intervertebral joint is decreased, the load through the plate increases. In the case of denucleation, the plate has to bear 100% of the load. In such cases, the load-sharing character of the spinal column is dramatically changed, or even lost [17].

The trabecular centrum is the dominant structural component of the vertebral body, while the shell accounts for only 10% of vertebral strength [24]. The importance of cancellous bone for load sharing in a human lumbar spinal segment has been shown using a finite element approach [20]. Increasing bone mineral density of cancellous bone resulted in an increase in the load that is passed through the disc and the vertebral bodies [20]. Moreover, cancellous bone density seems to influence the initial stability of different types of lumbar spine osteosyntheses to a great extent [19].

Two conclusions can be drawn from the results mentioned above. First, the spine is a load-sharing construction and load sharing can be approximated by surgical reconstruction to the physiological state, even after extensive destruction [2, 7, 10, 16, 17]. Second, cancellous bone of the vertebral body seems to play an important role for load sharing of the intact segment [20, 24]. However, the influence of cancellous bone density on load sharing in a surgically altered segment has, to the authors' knowledge, not yet been investigated. Additional knowledge concerning the distribution of forces within the stabilised spine depending on cancellous bone density could be helpful for performing spinal surgery in, for example, patients suffering from osteoporosis. Such information could be used to improve the shape and positioning of spinal devices. Furthermore, it could be helpful for better reconstruction of spinal biomechanics following surgery.

Thus, the aim of the current study was first to compare load sharing in compression in an intact and surgically treated lumbar spine motion segment L3/4. Second, the influence of cancellous bone density on load sharing in both models was defined.

In studying the influence of different parameters on spinal biomechanics, the finite element method can be a powerful tool [8, 9, 10, 11, 12, 13, 17, 18, 19, 20]. However, as Goel and Gilbertson pointed out [9], although the finite element method is increasingly able to simulate a variety of clinical situations in a realistic manner, a great deal of work must precede this (especially experimental validation). Thus, a biomechanically validated finite element model was used to examine the influence of cancellous bone density on load sharing in an intact and surgically altered human lumbar spinal segment.

Materials and methods

Biomechanical study

Six cadaveric lumbar spine segments (three segments L2/3 and three segments L4/5) were taken from 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 vertebra 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, Wehrheim, Germany) for rigid fixation of the motion segment.

The specimens were positioned horizontally in a tension/torsion testing machine (MMT 0005, Schenk, Germany) for axial compression (Fig. 1). Load was applied normal to the mid-disc plane. The polymethylmethacrylate was rigidly fixed in the testing apparatus, and the specimens were held in a saline solution during the testing [1, 6].

Measurements were first made on the intact motion segment. Following this, a bilateral laminectomy, resection of the flave ligament and bilateral resection of the medial parts of the facet joints were performed. Via a bilateral incision of the posterior longitudinal ligament and the posterior parts of the anulus fibrosus, the nu-

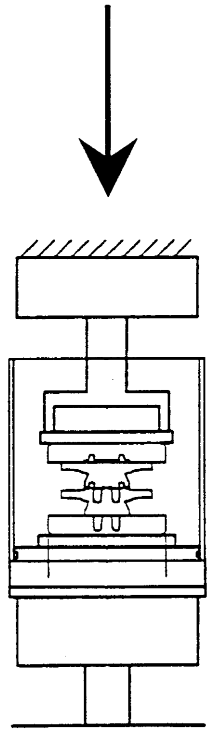

Fig. 1 Line drawing of the testing machine with a specimen

Table 1 Material properties of the bone, the disc, and the implants, taken from the literature [13, 15, 18, 22]

	E modulus (MPa)	Poisson's ratio
Shell, endplate	12,000	0.3
Cancellous bone	25	0.2
Pedicle	3500	0.25
Nucleus pulposus	1	0.499
Anulus fibres	450	0.30
Anulus substance	4.2	0.45
Implants	110,000	0.36

Table 2 Material properties of the ligaments, taken from the literature [13]

	E-modulus (MPa)		Area (mm^2)
Ant. Long. lig.	7.8 (<12%)	20 (>12%)	63.7
Post. Long. lig.	10 (<11%)	20 (>11%)	20
Flav. lig.	15 (<6.2%)	19 (>6.2%)	40
Caps. lig.	7.5 (<25%)	33 (>25%)	30
Interspin. lig.	10 (<14%)	12 (>14%)	40
Supraspin. lig.	8 (<20%)	15 (>20%)	30

cleus pulposus was removed completely. However, the lateral and anterior parts of the anulus were left intact. Autologous bone chips were made from the resected lamina and placed in the anterior one-third of the disc space. The internal stabilisation system (MOSS–MIAMI, consisting of titanium surgical mesh combined with a posterior screw/rod construct; DePuy International Ltd, Leeds, UK) was then installed on the motion segment. Two titanium surgical cages were sized and placed into the distracted disc space in tight contact with the adjacent vertebra. There was a distance of approximately 1 mm between them, and a distance of 2–3 mm between the posterior edge of the cages and the posterior wall of the vertebral bodies [14]. The posterior instrumentation was sized and implanted according to surgical practice and technique along the axis of the pedicles and was then connected by a rod on each side. Finally, the motion segment was placed in compression [14].

Biomechanical testing was performed using 600 N in axial compression [1, 6]. A compressive load was applied at the midpoint of the vertebral body of the top mounted vertebra (L2 or L4, respectively). The load was applied within 10 s with constant speed. Displacement was measured using an electronic displacement transducer, integrated in the bottom area of the testing machine. Hysteresis curves were plotted and used for calculating stiffness. The testing cycle was repeated three times, with the third cycle used for measurement and statistical analysis. Mean stiffness and standard deviation were measured for both the intact and the surgically repaired specimen.

Finite element analysis

The finite element programme ANSYS 5.4 (Swanson Analysis Systems, Inc., Houston, Tex., USA) was used to create the mathematical model of the lumbar spinal motion segment L3/4.

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 this was then interpreted as an anisotropic, nonlinear finite element model, based on the work done by Goel et al. [13] and Shirazi-Adl et al. [22]. The thickness of the endplates and cortical shell were set to 0.25 and 0.4 mm respectively [23]. Young's modulus of cancellous bone has a wide range – from 25 to 100 MPa, according to the literature [24]. The value of 25 MPa was used, as it was felt to approximate the cancellous bone in the samples of our patients with an average age of 64.8 years [24]. The structure of the anulus fibrosus was simulated by a ground substance embedded with 768 fibres at alternating orientations of 30° [13]. The nucleus pulposus was specified as an incompressible structure [13]. The two facet joints were simulated as gap elements, with all essential ligaments added to the bony structures, using material properties from the literature [13, 15, 18, 22], (Table 1, Table 2). The finite element model of the intact L3/4 motion segment (Fig. 2) consisted of 7433 single elements.

As the next step, the posterior elements and the inner parts of the disc were removed from the intact model to simulate the posterior decompression and discectomy, as performed in the biomechanical part of the study. Internal stabilisation hardware and arthrodesis bone were added to the defect segment model (Fig. 3). Two trapezoidal cages of titanium surgical mesh (11.2–12.3×16 mm), as described by Harms [14], were placed between the vertebral bodies. The posterior edges of these cages were placed 1 mm apart, 2.5 mm from the posterior wall of the vertebral bodies. The endplates of the vertebral bodies were not violated. Two pedicle screws (6×50 mm) were inserted through the axis of the pedicle with a convergence angle of 7° and with the rostral caudal angle of the pedicle, and connected by two rods (5×40 mm). The screw-bone interface was modelled by a coupling of X, Y, Z degrees of freedom with fine net, such that a tight interface was achieved. The surgically treated model (Fig. 3) consisted of 7100 elements.

Finally, three different E moduli (10, 25, 100 MPa) for cancellous bone density were used in both the intact and the surgically treated finite element models. These models were called intact FEM 10, 25, 100 and surgically altered FEM 10, 25, 100, and were used to predict load sharing in axial compression using 600 N normal to mid-disc height.

All FEMs were loaded as follows: the lower vertebra (L4) was not allowed to move in any direction. With the mid-disc plane ori-

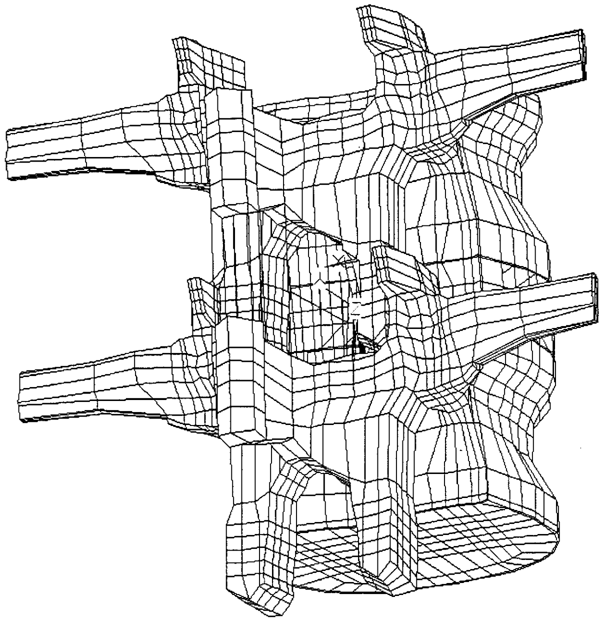

Fig. 2 Finite element model of intact L3/4

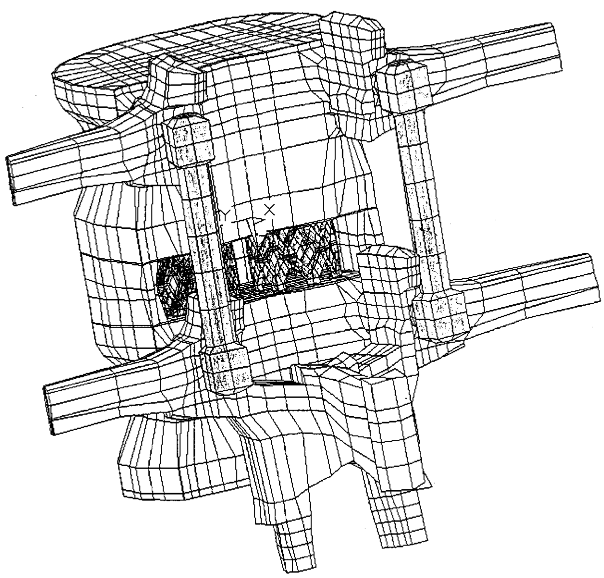

Fig. 3 Finite element model of surgically treated L3/4 (implants *shaded grey*)

ented horizontally, a compressive load of 600 N was applied to the upper vertebra (L3) in such a way that all nodes of the uppermost plane of L3 were loaded

Results

Biomechanical validation

Mean stiffness of the intact specimens in the biomechanical study was 654 (±220) N/mm for compression. There were no noteworthy differences between L2/3 versus L4/5 lumbar segments. The intact FEM 25 predicted stiffness of 699 N/mm.

Mean stiffness of the surgically treated specimen in the biomechanical part of the study was 895 (±313) N/mm for compression. The surgically altered FEM 25 predicted stiffness of 1134 N/mm (Table 3).

Note the similarity between the cadaver biomechanical measurements and the predictions made from the finite element model. The predictions of the finite element model are within one standard deviation of the mean stiffness as analysed by the biomechanical study. Thus, the finite element model was used to analyse load sharing in axial compression in an intact and a surgically treated human lumbar spine segment.

Table 3 Results of biomechanical and finite element analysis: stiffness of the intact and surgically treated motion segment (N/mm)

	Biomechanical analysis	Finite element analysis
Intact segment	654 (±220)	699
Surgically repaired segment	895 (±313)	1134

Load sharing in an intact motion segment

Using the intact FEM 10 for analysis of the load-sharing concept, the load through the anterior part was 86%. The load passed through the vertebral bodies and the disc increased to 89% using FEM 25 and to 91% using FEM 100 (Fig. 4).

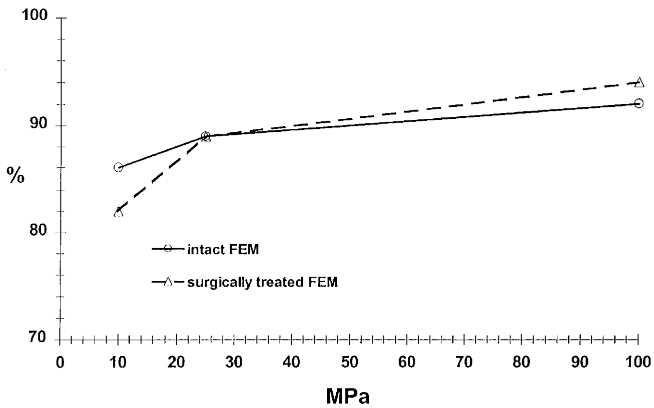

Fig. 4 Graph demonstrating load sharing in an intact (*black line*) and surgically repaired (*dotted line*) L3/4 motion segment, according to cancellous bone density. For each bone mineral density, given in megapascals along the *X* axis, the *Y* axis gives the percentage of compressive load that is passed through the anterior aspect (disc and vertebral bodies), represented by the area below the line. The area above this line represents the percentage load that passes through those posterior elements that are able to carry compressive loads (i.e. the facets)

Load sharing in a surgically altered motion segment

Using the surgically altered FEM 10 for analysis of the load-sharing concept, the load through the anterior part was 82%. The load passed through the vertebral bodies and the disc increased to 89% using FEM 25 and to 92% using FEM 100 (Fig. 4).

Discussion

The current study was performed to investigate the influence of bone mineral density of cancellous bone within the vertebral bodies on load sharing in intact and surgically stabilised human lumbar spine. A validated finite element model of an intact and surgically treated human lumbar spinal segment L3/4 was used.

We can conclude from a variety of previous investigations that lumbar spinal load sharing is performed by passing 80–96% [2, 7, 8, 10, 12, 16] of the applied load through the anterior part of the spine and the remainder through the posterior elements. These results are comparable to the findings of our finite element analysis. Thus, it seems to be clear how much of the applied load passes through the anterior part of the spinal column, i.e. 80–96% of compressive forces.

These results, however, do not apply for every loading condition. Load sharing very much depends on posture. It has been shown that, in compression, the facets become load bearing in extension, whereas in flexion they are not [2, 3, 7]. The facets carry 10–40% of an applied compression in extension [2, 7]. Disc space narrowing also has a significant influence on load sharing of the human lumbar spine. Narrowing of the disc space by 1 mm through application of a compression force of 2000 N for about 2 h resulted in a significant increase in load carried by the facets when compared to the intact state [7]. Thus, load sharing could also be influenced by the loading history [7].

To predict the biomechanical changes caused by stabilisation procedures to an intact spinal segment, a few finite element models have been described [10, 11, 12, 17]. It was found that there is an increase in stiffness of the stabilised segments when compared to the intact ones. This is in substantial agreement with the results of our current study. Moreover, Goel et al. [10], in an analysis of load sharing in human lumbar spine using a finite element approach, found that 96% of an applied compression force is passed through the disc in the intact spine, while 80% of compression load passes through an interbody bone graft, that means through the anterior part of the lumbar spine. If, however, a posterior screw-plate osteosynthesis is used without interbody bone graft, the load through the plate increases to 40% of the applied compression force. Furthermore, Lim and Goel [17] found that, if the stiffness of the intervertebral joint (disc) is decreased, the load through the plate increases. In this case, the load-sharing character of the spinal column has been changed dramatically [17]. These latter results, in particular, are in agreement with the results presented here. Both investigations found increased load through the posterior elements as a result of changes to the anterior aspect of the spine: in the study by Lim and Goel [17], this was achieved through decreased stiffness of the disc, and in our study, through decreased stiffness of cancellous bone density in the vertebral body. Thus, the results of the current study agree with the results of former studies by others, as mentioned above.

Some simplifications are included in the finite element analysis presented in the current study. Due to a large amount of dense cortical and subcortical bone represented in the outer parts of the pedicles, the influence of a small amount of cancellous bone in their centre has been neglected. The pedicles were generated as solid bony structures with an E modulus of 3500 MPa [13], thus imitating the pedicle as a cylinder of bone with little cancellous bone in its centre [21]. According to the findings of Silva et al. [24], an E modulus of between 25 and 100 MPa should be appropriate for cancellous bone. However, a lower value (10 MPa) was also used in our study. This was done to investigate the influence of very low bone mineral density on spinal load sharing as it occurs, for example, in cases of metastatic destruction, which is of some clinical importance. Moreover, our investigation was limited to an analysis of load sharing in axial compression. Previous work with this finite element model has shown that the influence of cancellous bone density on the biomechanical behaviour of the segment is most pronounced in axial compression [19]. The biomechanical analysis in this study was performed using three L2/3 segments and three L4/5 segments. We chose only the L3/4 segment for constructing the finite element model, however, due to the precise description of the material properties of a human spinal segment L3/4 in the literature [13]. There were no noteworthy differences in stiffness between L2/3 and L4/5 segments in our biomechanical testing.

Although the results of the biomechanical and finite element study are not exactly the same, they are comparable because the results of the finite element analysis are always within one standard deviation of the results of the biomechanical experiment (Table 3). The minor differences can be easily explained by biological variability in the sample and the fact that the material properties are only approximate for the sample analysed. Thus, the finite element model based upon the models by Goel et al. [13] and Shirazi-Adl et al. [22], and modified by ourselves concerning the surgical approach and surgical implants, is reasonably validated by comparable cadaver analysis and is considered useful in making further predictions.

Surgical treatment of spinal disorders usually calls for destruction of the anterior or posterior parts of the spine or even both. Thus, spinal integrity and biomechanics are compromised. Kyphotic deformity, progressive listhesis or even scoliosis are well-known long-term complications following destruction of the posterior parts of the spine or laminectomy [5, 25, 26], unless precise and limited decompression [4] has been performed. These are classic examples of the consequences arising from deranged spinal biomechanics. For these reasons, reconstruction of spinal biomechanics should be a main goal in spinal surgery. The current study clearly demonstrates the importance of posterior elements for spinal load sharing of compression forces. Following complete removal of the posterior elements of the spine, the anterior column has to bear 100% of the applied force. Thus, the spinal load-sharing construction is destroyed after resection of the posterior parts of the spine; reconstruction is therefore a logical step. A pedicle-screw-rod system was used for posterior stabilisation in the current study. The resected parts of the disc were replaced by two pieces of titanium surgical mesh filled with cancellous bone. Thus, the posterior and the anterior part of the spine were reconstructed. Reconstruction of both the anterior and posterior parts of the spine restores the load-sharing concept. The spine again acts as a load-sharing construction, and the manner in which compression is shared between the anterior and posterior parts is close to that in the intact spine motion segment, as defined by the finite element analysis. The similarity in load sharing between normal and instrumented spines appears to depend on assumed bone density, and it may depend also on applied load, and loading history. The anatomical reconstruction of the spine as described above also mimics its physiological properties.

Despite these biomechanical advantages of the reconstruction of the spinal column, it remains unclear whether clinical outcome is improved by the biomechanically correct reconstruction of the spine. Obviously, reconstruction of the anterior and posterior parts of the spine seems to be correct from a biomechanical point of view. Conversely, this may not influence clinical outcome related to pain and neurological deficits. These parameters, however, are either not, or only partially, influenced by biomechanical properties.

Conclusion

Load sharing between the anterior and posterior parts of the human lumbar spine is influenced by the cancellous bone density of the vertebral bodies. The higher the bone mineral density, the higher the percentage of compressive load that is passed through the anterior part of the lumbar spinal column. Following posterior decompression (i.e. laminectomy and medial facetectomy) and discectomy, and thus destruction of the spinal load-sharing character, reconstruction of both the anterior and the posterior elements with the implants used in the study recreates the ability of the spine to act as a load sharing construction. In this case, load sharing is again influenced by bone mineral density.

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