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The effect of bone graft geometry on spinal fusion vertebral stresses

D. LACROIX[1], J. NOAILLY[1], G. SALO[2], E. CÀCERES[2], J.A. PLANELL[1]

[1]Bioengineering Research Center of Catalonia (CREBEC), Department of Material Sciences, Universitat Politècnica de Catalunya, Barcelona - Spain
[2]IMAS, Hospital del Mar, Barcelona - Spain

ABSTRACT: *Spinal fusion is a frequent surgical technique in which the success is uncertain due to post-operative changes in the biomechanics of the spine. Bone grafts are good candidates for disc and vertebra substitutes due to their similar bone properties and their good osteogenic properties. However, the effect of the anatomic harvest location of the bone graft on the load transfer is unknown. A physiologic three-dimensional (3D) finite element model of a lumbar spine was modified to model spinal fusion with a fixator and a bone graft. Bone grafts were taken either from the femur, the tibia, or from the fibula in a configuration of three or six fragments. The configurations were submitted to physiological loadings, and strain and stress distributions were calculated within the vertebrae, the fixator and the bone grafts. Quantitative differences were found from one type of bone graft to another. It was found that fibula bone grafts provided better stability by carrying a large part of the load. However, femoral and tibial bone grafts provided a more similar strain distribution within the vertebrae compared to the physiologic model. For tibial bone grafts, load transfer was found to be sensitive to the orientation used during the surgery. The use of a femoral bone graft to replace one vertebra and two intervertebral discs was found to give a better biomechanical function than using a tibial or fibula bone graft. This surgical technique is proposed to be beneficial in the case of severe spinal trauma providing good interface is obtained between the bone graft and the vertebrae.(Journal of Applied Biomaterials & Biomechanics 2006; 4: 135-42)*

KEY WORDS: *Spine, Fusion, Finite element analysis, Bone graft*

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INTRODUCTION

In spinal vertebral fusion, there is a plethora of choice of bone substitutes ranging from natural cadaveric bones to synthetic materials commonly known as cages. Due to their natural properties, bone grafts are a tempting solution. However, the choice of bone graft type, ie originating either from the femur, the tibia or the fibula, is still unsolved. In this study, a biomechanical analysis of the various anatomic locations of bone grafts was performed using finite element analysis.

Despite the availability of various mesh and cylindrical cages, a bone graft remains a simple and adequate alternative for anterior structural support in spinal surgery. Moreover, long-term studies have not indicated that cages are superior to bone grafts (1). Bone substitutes with cages can induce complications such as cage subsidence within the cartilage endplate leading to pain and mechanical instability.

Clinical results of anterior fusion in the short term are good, but are still debated regarding the long term (2). In a long-term follow-up study of lower lumbar fusion, accelerated degeneration of adjacent segments and segmental instability were reported in 45% of patients (3). Removal of a vertebra instead of only one disc is even more challenging. Results in the mid to long term indicate that fusion is successful in most cases, although lucency at the top and bottom of the graft can be found (4). Results are better in anterior fusion when the bone graft is combined with posterior segmental instrumentation (4, 5). A bone graft has the advantage of reducing stresses in the fixator (6). However, as expected, the substitution of one component for another changes the stress state within the other components of the spine and the load sharing within the spine (7). This can lead to long-term complications through remodeling.

The variability in clinical results in terms of patients, fusion methods and follow-up makes it diffi-

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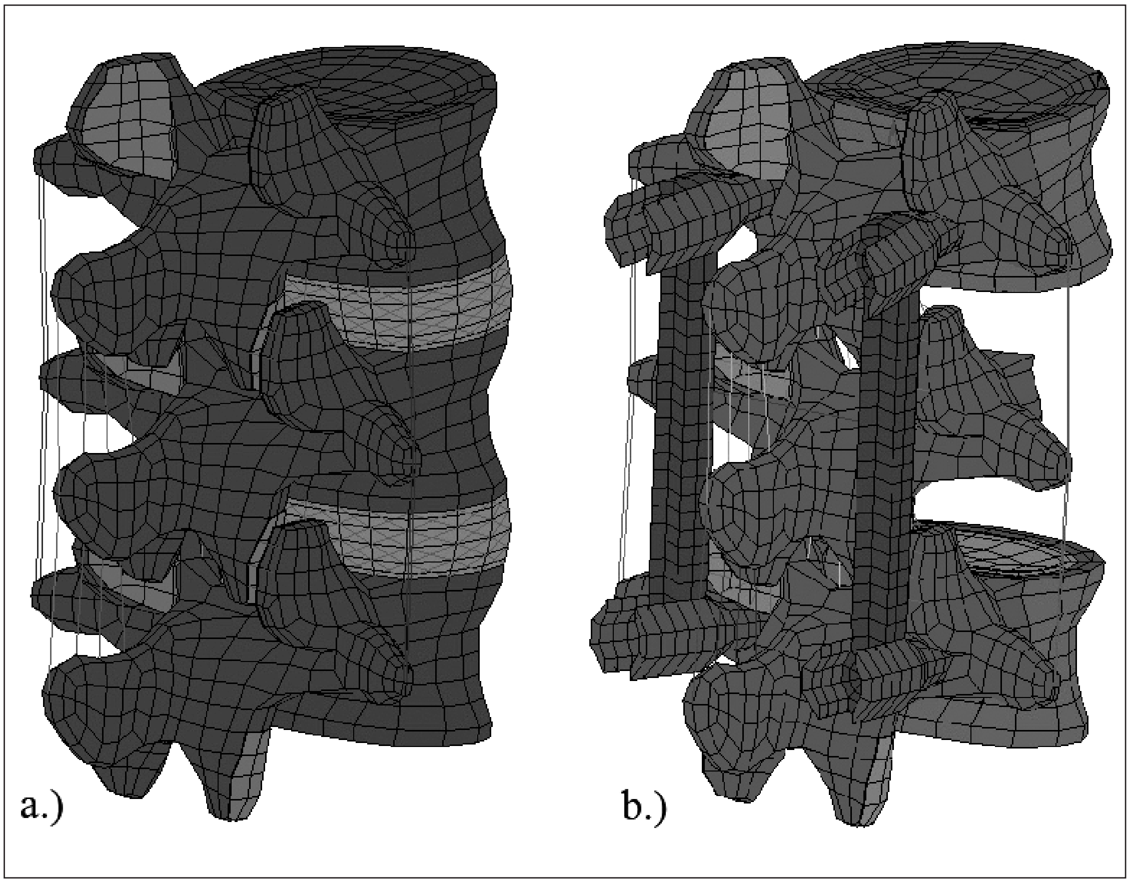

Fig. 1 - ***Finite element mesh of the L3-L5 lumbar vertebra a) in a physiologic model, b) with resection of L4 vertebra and fusion of L3 and L5 vertebrae through a spinal fixator.***

cult to assess which type of bone graft is better over another. The finite element method is a suitable approach for studying only a few parameters without altering the other variables. It allows the prediction of the influence of one parameter on others. Finite element studies of anterior interbody fusion have been performed previously, but have investigated only the substitution of a disc and not of a vertebra (7-9).

The objectives of this study were twofold: 1) to use the finite element method to investigate the effect of substituting one lumbar vertebra with a bone graft on the stresses within the adjacent vertebrae and; 2) to investigate the influence of bone graft anatomic location on the stresses within the adjacent vertebrae.

METHODS

A non-linear three-dimensional (3D) finite element model of a lumbar spine L3-L5 was used as a physiologic model (10). The model was taken from Smit (11) and modified to obtain an intact model where all spinal cartilages, ligaments and bony elements are represented (Fig. 1). In the physiologic model, cartilage endplates were modeled to cover the whole nucleus pulposus and spread over the annulus fibrosus up to between one-third and half the annulus thickness. Their thickness varied from 1 mm to the periphery to about 0.6 mm in the center (12, 13). Annulus fibrosus fibers were distributed into 20 concentric and mono directional fiber layers. The fiber section and volume fraction within each layer varied with the layer radial location. The fiber orientation varied in the radial direction in a crisscross pattern from 62-45° (14). Articular facets were represented by 3D hexahedral elements 2 mm thick (15, 16). The contact area was interpolated by Coon surfaces. Capsular ligaments were modeled using truss elements, which formed a ring around the articular contact and bound together the inferior and superior facets (17, 18). The six other spinal ligaments were modeled with truss elements and their sectional areas were taken from the literature (18).

All stress-strain relations were calculated for large displacements and large strains. Table I summarizes the constitutive laws and material parameter values used for the tissues modeled. The non-linear behavior of the supraspinous, interspinous, intertransverse flavum

and capsular ligaments were taken from the literature (15, 19).
The model was modified with the insertion of an internal fixator (Sherpa® Spine System, Surgival SA, Spain) between L3 and L5 with the removal of the two intervertebral discs and the L4 vertebra (Fig. 1). Contact between the screws and the bone was assumed as perfect with no motion. The fixator was made of titanium with a Young's modulus of 110 GPa and a Poisson's ratio of 0.3.
CT scans of a femur, a tibia and a fibula from the same patient were performed. Fragments of each bone were reconstructed and inserted within the model. Four bone graft configurations were investigated: one femur fragment, one tibial fragment, three fibula fragments, and six fibula fragments (Fig. 2). The bone grafts were considered as homogeneous and anisotropic materials with properties defined in Table I.
Following Chen et al (20), a solid fusion was modeled, ie able to transmit loads in compression and tension, to represent a well advanced post-operative stage to study the influence of each bone graft in the long term. In the clinical community, it is debated whether to keep as much as possible of the cartilage endplate or only part of it: the cartilage endplate has the advantage of providing support, but with reduced vascularisation that can be detrimental for bony union (2). Therefore, in this study the influence of the mechanical support of the endplate was investigated by modeling either 1) a full thickness cartilage endplate (around 1mm); or 2) half the thickness of a cartilage endplate; or 3) no cartilage endplate.
Four types of loadings were applied on L3 vertebra: compression of 1000 N, and moments in flexion, and extension and rotation of 15 Nm to represent physiological loading. Fixed displacements were imposed on the distal part of L5. Static simulations were carried out using MSC MARC 2001 (MSC Software).

RESULTS

The use of a spinal fixator significantly increased the stiffness of the whole structure. The stiffness increase in flexion was 116% with a femoral bone graft. In all load cases, the fixator was well below the titanium fatigue limit with a maximum von-Mises stress of 110 MPa with a femoral bone graft in the case of flexion. Therefore, no complications specifically with the internal fixator were foreseen.
Displacements of the L4 vertebra were calculated for all types of bone graft and loadings (Fig. 3). Results indicated that flexion is the most highly stressed loading for the fixator and the vertebrae. In flexion, compared to the femur (that has a stiffness of 3900 Nmm/mm), the tibia was 15% stiffer, the fibula with three fragments was 28% stiffer and the fibula with

TABLE I - MATERIAL PROPERTIES USED IN THIS STUDY

Material	Constitutive laws	E (MPa)*	υ	G (MPa)
Trabecular bone	Orthotropic elastic (25, 26)	140, 140, 250	0.45, 0.31, 0.3	38, 77, 77
Cortical bone	Orthotropic elastic (26, 27)	8000, 8000, 12000	0.4, 0.35, 0.3	2000, 2400, 2400
Bony endplate	Isotropic elastic (25)	1000	0.3	-
Bony posterior elements	Isotropic elastic (28)	3500	0.3	-
Cartilage endplate	Isotropic elastic(25, 27, 29)	24	0.4	-
Facet cartilage	Tension isotropic elastic	11 (15)	0.2 (30)	-
	Compression non-linear elastic	From 11 at 0% strain to 3500 at 0.7% strain	From 0.2 at 0% strain to 0.4 at 0.7% strain (30)	-
Ligaments	Non-linear elastic	Experimental data (31-33)		
Annulus fibrosus fibers	Non-linear elastic	Collagen I: Experimental data (15)		
Annulus fibrosus matrix	Neo-Hookean (34)	μ = 0.5MPa		
Nucleus pulposus	Mooney-Rivlin (11)	C10=0.12MPa, C01=0.03MPa		
Fixator	Isotropic elastic	110000	0.3	-
Bone graft	Anisotropic elastic (35)	11900, 11900, 19900	0.42, 0.23, 0.23	4000,5200, 5200

* The Young's moduli are presented, respectively, in the 11, 22 and 33 directions. The Poisson's ratios and the Coulomb's moduli are given, respectively, in the 12, 23 and 31 directions; 1: coronal plane horizontal direction, 2: sagittal plane horizontal direction, 3: axial direction

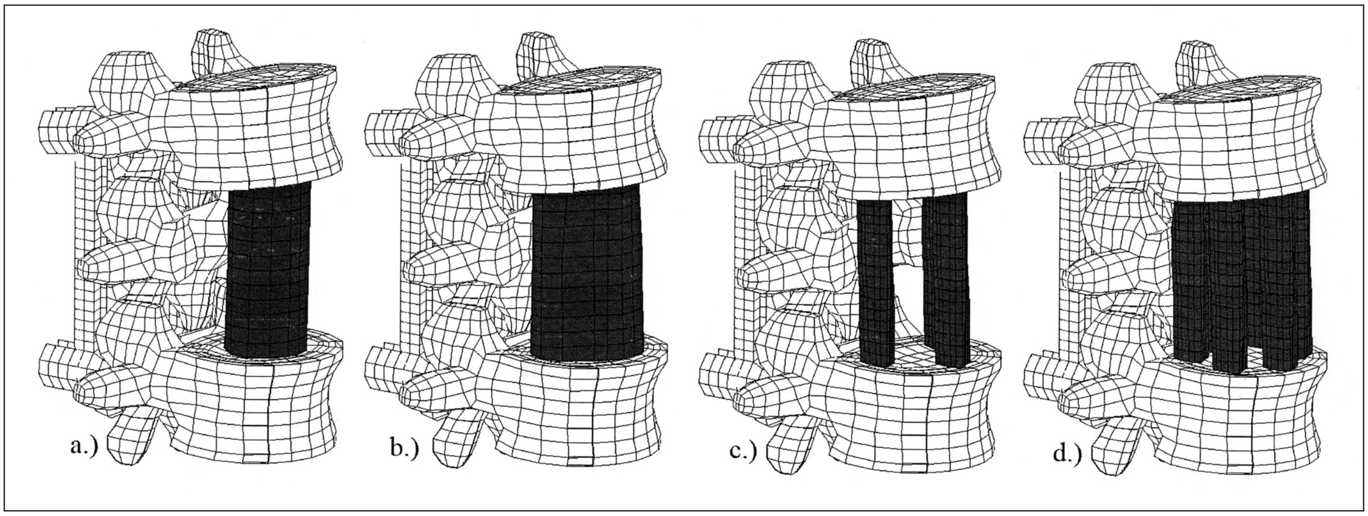

Fig. 2 - ***Finite element models of the L3-L5 lumbar vertebrae with a) a femoral, b) a tibial, c) three fibula fragments, d) six fragments of fibula bones.***

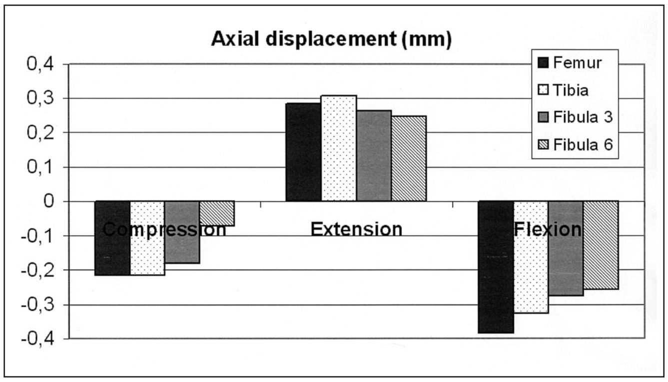

Fig. 3 - ***Axial displacement of L4 vertebra in the cases of compression, extension and flexion and for the four bone graft types.***

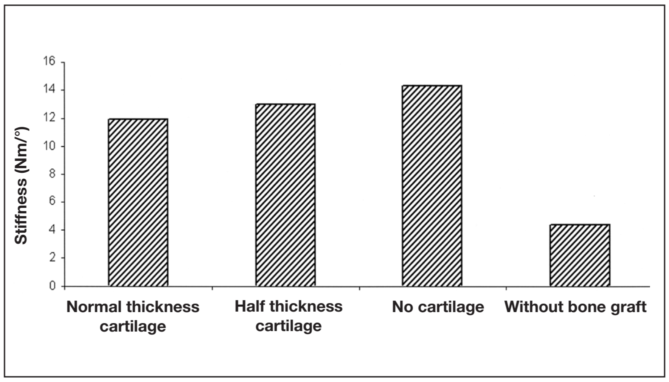

Fig. 4 - ***Stiffness values in flexion for three thicknesses of cartilage endplate with a tibial bone graft: normal thickness cartilage, half thickness cartilage, no cartilage endplate, and without bone graft.***

six fragments was 33% stiffer. In all loadings, the use of six fibula fragments gave the stiffest configuration. It should be pointed out that bone graft use compared to a physiologic model increased the overall stiffness of the system by almost one order of magnitude.

When the thickness of the cartilage endplate was varied, it was found that the stiffness increased as the thickness decreased, demonstrating the load "softening" effect of the endplate (as shown in flexion for the tibial graft in Figure 4). The change in thickness of the cartilage endplate was also followed by a change in the load transfer with a reduction in the role of the load transfer of the fixator, as the thickness of the cartilage endplate decreased. Stiffness was greatly increased with the bone graft as compared with the situation without the graft (Fig. 4).

When comparing the major principal stress distribution obtained in a physiological model with a spinal fusion model, it was found that stresses within the vertebrae and the processes were understressed when a fixator is used (Fig. 5). The tibia and femoral bone grafts were largely understressed, whereas fibular bone fragments were more highly stressed. Moreover, fragments on the anterior side were more stressed than on the posterior side. Cross-sections through L5 showing the major principal strains indicated that the strain distribution with a bone graft made of femur or tibia was more similar to the strain distribution in a physiologic model than with fibula fragments (Fig. 6).

Therefore, since the configurations with the fibula gave the highest stress concentrations within the vertebrae, it could be expected to correspond to the sit-

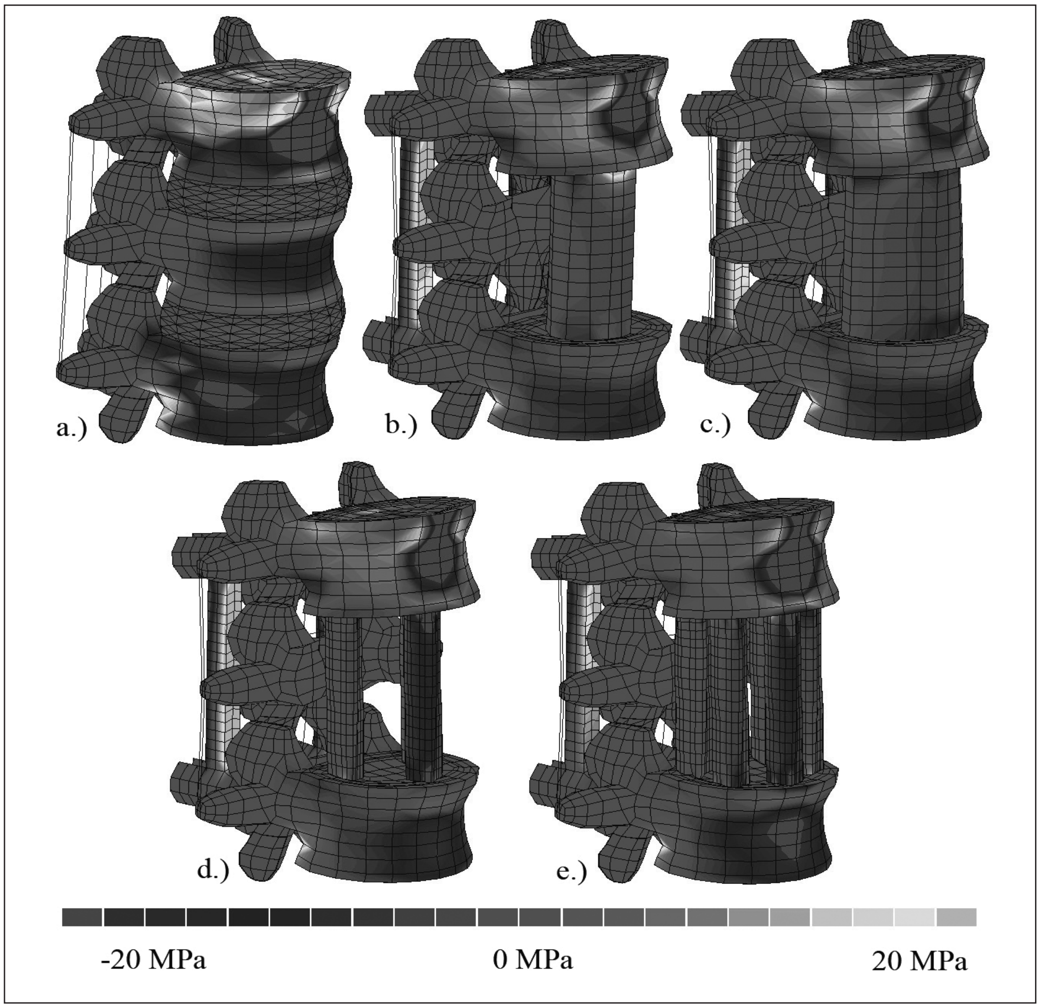

Fig. 5 - ***Principal stress distributions under flexion in a lumbar segment a) in a physiologic model, b) fused with a femoral bone graft, c) fused with a tibial bone graft, d) fused with three fibula bone grafts, e) fused with six fibula bone grafts.***

uation where bone remodeling is at risk, which could lead to further complications. In the case of the tibia, it was found that the asymmetric geometry of the tibia was more prone to rotational instability due to its geometry. When rotating the tibia by 90°, major principal stresses within the vertebra under compression were higher (Fig. 7) indicating that the positioning of the tibia can be very important for the load transfer.

DISCUSSION

A finite element analysis of a spinal fusion of two intervertebral discs and one lumbar vertebra replaced by a bone graft was performed. It was found that the use of a bone graft in combination with an internal fixator drastically changes the physiologic load transfer. Moreover, the use of bone grafts of various geometry and anatomical location can lead to differences in the biomechanics of the spine. The stiffest grafts were found to be the fibula fragments in a combination of three or six. This correlated well with results obtained by Siff et al (21): in a comparison between femoral ring *vs* fibular strut allografts in anterior lumbar fusion, they found that a fibular strut allograft creates a more stable and rigid construct, although not statistically significant. However, our study indicates that the use of fibula fragments induces a higher change in stresses and strains within the vertebra. Therefore, in the long term, the clinical outcome could be negatively influenced by the high modification of the stress-strain distribution obtained by this type of bone graft. On the contrary, bone grafts made of tibia and femur best preserve the strain-stress distributions within the vertebrae. Nonetheless, it was found that due to its geometry, a tibial bone graft can induce dissymmetry in biomechanical behavior; and therefore, is more sensitive to the position initially given by the surgeon. The asymmetry can be detrimental in the load transfer process and can create some instability. Therefore, a femoral bone graft seems to be more suitable.
As part of a finite element study, a number of limita-

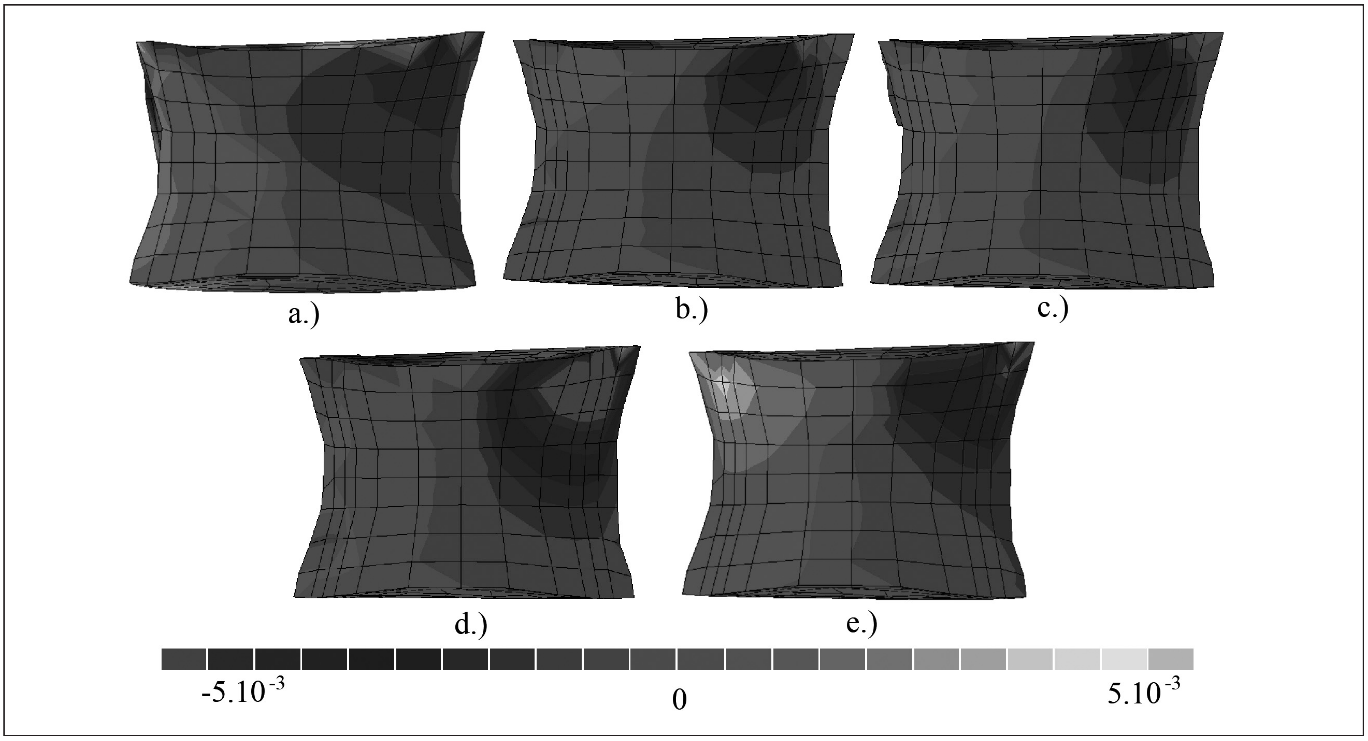

Fig. 6 - ***Principal strains in the L5 cross section vertebra. a) physiologic model, b) femoral bone graft, c) tibial bone graft, d) with three fibula segments, e) with six fibula segments.***

tions must be underlined. The 3D physiological model that was modified to insert an internal fixator was based on only one patient. This model has defined material properties, contact conditions and geometry that can vary from one patient to another. However, Smit previously validated this model (11). In this study, the model was used to assess the effect of different bone allografts; and therefore, it is suitable as a comparison study between models where one or various parameters have been changed. The results should be interpreted mainly as a comparison between the models studied. The fixator was defined completely anchored within the bone and with no pretension. Although pretension can modify the stress state within the bone graft (22), this study is mainly comparative; and therefore, the results obtained in the comparison of bone graft should not differ. This is particularly true due to the contact condition between the bone graft and the vertebrae where complete bone was assumed to model an effective fusion.

The stress analysis performed in this study gave a quantitative indication on stresses within an internal fixator. It was found that stresses within the fixator were well below the yield strength and fatigue limit in all load cases. Therefore, one should not only design a fixator to reduce stresses in the fixator, but to see the influence of the fixator on the other spinal components. In all load cases, the use of a fixator with a bone graft considerably changes the stress-strain distributions, which in the long term would involve an adaptative process that could be simulated using a remodeling algorithm (23, 24).

The thickness of the cartilage endplate was shown to play an important role in the load transfer to act as a damping of the structure. In a femoral bone graft, the thickness of the cartilage endplate can increase the stiffness in flexion of the lumbar spine by 14%. Therefore, biomechanically, it is recommended to keep the endplate as much as possible. However, a compromise must be achieved with biological considerations to obtain sufficient angiogenesis for bone healing.

CONCLUSIONS

This study has demonstrated the possibility of using the finite element method as a clinical tool in the bio-

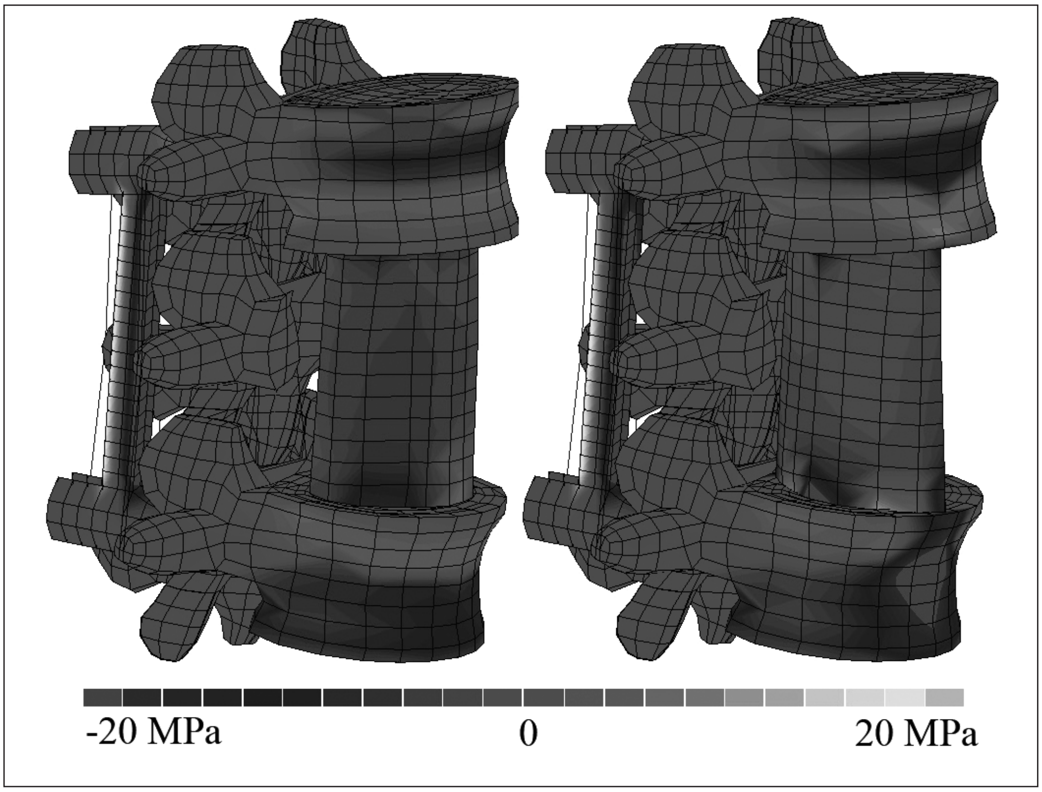

Fig. 7 - ***Principal stress distributions in a fused lumbar segment under compression with a tibial bone graft with the major axis aligned with a) the coronal plane, b) the sagittal plane.***

mechanical assessment of the choice of one type of bone graft over another. This type of study has the advantage of changing only a few controlled parameters to discard many other variables that are difficult to control experimentally or clinically such as donor properties (stiffness, size, age, osteoporosis state, etc), control of bone graft height and size, as well as contact conditions between the bone graft and the vertebrae.

The use of a femoral bone graft to replace one vertebra and two intervertebral discs was found to be better in terms of load transfer than using a tibial or fibula bone graft. This surgical technique is proposed to be beneficial in the case of severe spinal trauma providing good interface is obtained between the bone graft and the adjacent vertebrae.

Address for correspondence:
Damien Lacroix, PhD
Universitat Politècnica de Catalunya
Department of Material Sciences - ETSEIB
Diagonal 647
08028 Barcelona - Spain
damien.lacroix@upc.edu

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