Biomechanical Performance of Anterior Grafts in Lumbar Spine Surgery. A Comparative Finite-element Analysis*

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Purpose. To study the biomechanical performance of various allografts and the effects of endplate treatment on a lumbar corporectomy model.
Methods. A modified non-lineal tri-dimensional finite-element model of the lumbar spine was used, to which a set of transpedicular instruments was adapted. By means of a finite-element analysis, modeling was carried out of diaphyseal fragments of the femur, the tibia and the fibula. Four configurations were analyzed: with one femur, with one tibia, with three fibulas and with six fibulas. Four surfaces were evaluated that gave support to the graft according to the resection of the various components. Compression loads of 1,000 N were applied, as well as flexion, extension and rotation of 15 Nm respectively. The stresses and displacements caused were calculated.
Results. Full cartilage and subchondral bone resection is the configuration that least disrupts stresses within the adjacent vertebrae whereas the use of fibular fragments causes the greatest disruption. The use of the tibial bone gives rise to an asymmetry in the displacement area because of the shape of the said graft. The femur does not bring about a significant disruption of stresses in the adjacent vertebrae thereby constituting a more physiological construct.
Conclusions. Preservation of the endplate's cortical bone does not lead to any biomechanical advantage in the reconstruction of the anterior spine. Femoral allografts are the most appropriate ones to replace the vertebral body, compared with the tibia or the fibula.

Key words: *biomechanics, lumbar fusion, finite-element analysis, bone allograft.*

Comportamiento biomecánico del injerto anterior en la cirugía del raquis lumbar. Estudio comparativo mediante un modelo de elementos finitos*

Objetivo. Investigar el comportamiento biomecánico de diversos aloinjertos y el efecto del tratamiento del platillo vertebral en un modelo de corporectomía lumbar.
Método. Se utiliza un modelo modificado no lineal de elementos finitos en tres dimensiones de la columna lumbar al que se adaptó un instrumental transpedicular. Se modelaron por elementos finitos un fragmento diafisario de fémur, uno de tibia y uno de peroné. Se investigaron cuatro configuraciones: con fémur, con tibia, con tres peronés y con seis peronés. Se evaluaron cuatro superficies sobre las cuales se sustentaba el injerto en función de la resección de los distintos componentes. Se aplicaron fuerzas de compresión de 1.000 N, flexión, extensión y rotación de 15 Nm respectivamente. Se calcularon las tensiones y desplazamientos generados.
Resultados. La resección completa de cartílago y hueso subcondral es la configuración que menos altera las tensiones dentro de las vértebras adyacentes. El uso de fragmentos de peroné modifica en mayor medida las tensiones en las vértebras adyacentes. El uso de tibia genera una asimetría en los campos de desplazamiento debido a la forma de dicho injerto. Los resultados con fémur modifican en

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menor medida los estreses en las vértebras adyacentes, configurando un montaje más fisiológico.
Conclusiones. La preservación del hueso cortical del platillo vertebral no ofrece ninguna ventaja biomecánica en la reconstrucción de la columna anterior. El aloinjerto de fémur es el más adecuado para la sustitución del cuerpo vertebral, comparado con tibia y peroné.

Palabras clave: *biomecánica, fusión lumbar, análisis por elementos finitos, aloinjerto óseo.*

Many pathological conditions (such as tumors, fractures, infections and others) cause destruction of the vertebral body during their evolution. Spondylectomy is the technique of choice in these cases, but it is well-known that the vertebral body biomechanically supports approximately 80% of the load of the lumbar spine[1]. When the vertebral body is absent posterior instrumentation systems are subjected to excessive load that causes their failure, if their placement is not supplemented by anterior support at the interbody site[1,2].

To prevent this situation, the possibilities of providing anterior support in the interbody area are several, such as: the use of biomaterials (coralline hydroxyapatite), titanium cages, titanium meshes, cement (polymethylacrylate) and grafts, which have the biological advantage of achieving osteointegration. As to grafts, these may be cortico-cancellous or cortical, and in spite of the availability of several types and designs of cages and meshes, bone grafts continue to be an appropriate option as anterior support for spine column surgery. In spite of the fact that there are biomechanical studies that assess the behavior of cages placed to substitute the vertebral body, there are no clinical studies that indicate that these are the best structural graft in the long term[3].

According to several authors, anterior fusion has a good clinical outcome. X-ray results in the long term indicate that fusion is achieved in a high percentage of cases in spite of the occasional existence of radiolucencies in the contact zone of vertebral end-plates[4,5].

To stabilize the spine, it is indispensable to place posterior instrumentation while waiting for bone consolidation to take place[5,6]. It is well known that an anterior bone graft is capable of reducing stresses in the posterior fixator, as described by Atienza et al[7], both with flexion bars (82% with cortical grafts) as with flexion transpedicle screws (78% with cortical grafts), as also by an increase of rigidity to flexion (234% with cortical grafts). These authors conclude that in the case of spondylectomy an anterior graft must be incorporated, since the stresses achieved by the screws and bars in fixation systems are close to the yield strength of the titanium alloy (795 MPa)[7]. However, substitution of the vertebral body by a graft changes the stresses in adjacent vertebra and load distribution in the spine[8].

The variability of clinical results due to patients, fusion methods and follow-up markers make it difficult to assess the best type of graft for anterior substitution of a vertebral body. The finite elements method is an appropriate method for studying some parameters of biomechanical behavior of these anterior grafts as they control other variables that cause confusion. This allows prediction of the influence of one parameter on the others. Finite elements studies of anterior fusion have been carried out before, but they have focused on the analysis of substitution of the intervertebral disc, not of a complete vertebra[8-10].

In spite of the fact that, as we have already mentioned, it is known that good results have been seen with structural allograft in spine surgery, the best choice of type of interbody cortical graft has not yet been established. The aim of this work is to investigate the biomechanical behavior of different allografts (tibia, femur and fibula) placed anteriorly to substitute the vertebral body, and the effect of treatment of the vertebral end-plate on load transmission in a lumbar spondylectomy model.

MATERIALS AND METHODS

Finite Elements Model Geometry

A finite elements non-linear three-dimensional model of segments L3-L5 of the lumbar spine was used as a physiological model (Noailly et al[11,12]). The model was created and validated originally by Smit[13,14] and modified to obtain a better physiological model with all the cartilage components of the vertebral body, all the ligaments and all the bone elements (Fig. 1). The overall geometry was obtained from a CAT scan of the lumbar spine of a healthy 44 year old male, adapted to the sagittal plane and carried out with 2 mm sections subsequently reconstructed.

The dimensions and shape of each component were established based on the literature and anatomical data, and the bone was defined as isotopic. In the physiological model, the cartilages of the vertebral end-plate were modeled covering all the *nucleus pulposus* and extending up to the *annulus fibrosus*, from 1/3 to ? of this. The thickness of the vertebral end-plates varied from approximately 1 mm in the periphery to 0.6 mm in the center[15,16]. The vertebral body was formed by a center of cancellous bone surrounded by a layer of cortical bone. Proximally and distally it was formed by the terminal end-plate, which in this model had a thickness that was the average of the values collected on the sagittal plane. The values for the terminal end-plates were calculated assuming that the ratio of the average thickness

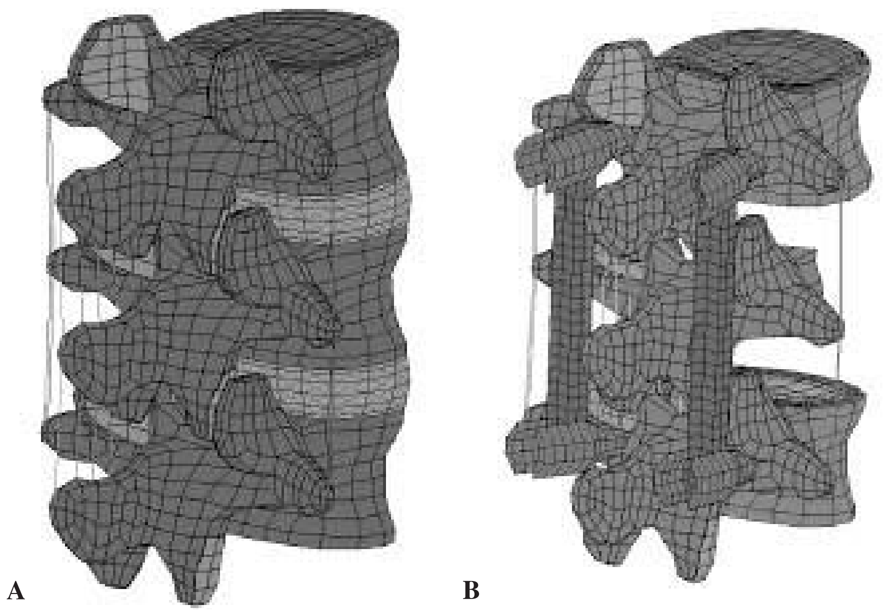

Figure 1. Mesh of finite elements of the lumbar spine L3-L5 segment in (A) a physiological model, (B) with the resection of the L4 vertebral body, discs L3-L4 and L4-L5 and with the adapted transpedicle fixator.

of the cortical and the terminal plates was 1.67 for one vertebra[17].

The terminal end-plate was designed with subchondral cancellous bone, a thin layer of cortical bone and two layers of cartilage on the end-plate (one deep and one superficial). The fibers of the *annulus fibrosus* were distributed between 20 concentric monodirectional layers. Sections of the fibers and the volume within each layer varied with the location of this in relation to its depth. The orientation of the fibers varied due to radial location, in an intercrossing pattern of 62° to 45°[18]. The joint facets were represented by hexahedral three-dimensional solid elements with a thickness of 2 mm[19,20]. The capsule ligaments were modeled using truss type elements, which formed a ring around the joint surface and surrounded the upper and lower facet[21,22]. The section area of the capsular ligaments varied according to the experimental measurement of their transverse and sagittal sections.

The other 6 vertebral ligaments were modeled with three-dimensional uniaxial truss type elements and there section areas adapted to data in the literature[22]. In this way, for one vertebra and from one vertebra to the next one, there are 98 parallel truss elements for the capsular ligament, 8 for the intertransverse ligament, 1 for each supraspinous ligament, 3 for each interspinous ligament and 3 for the yellow ligament. From the practical point of view, the anterior and posterior longitudinal ligaments, only possess biomechanical effects on the disc, since they are strongly attached to the *annulus fibrosus* and do not carry out any biomechanical action on the vertebral body. Finally, the model was made up of 6,158 isoparametric elements with 8 or 6 nodes, 215 unidirectional elements with 2 nodes and 6,375 nodes.

Transpedicle Vertebral Instrumentation

The model was modified with the insertion of a transpedicle instrumentation device for both L3 and L5 pedicles (Sherpa® Spine System, Surgival SA, Spain) and the removal of the 2 intervertebral discs and the vertebral body of L4 (Fig. 1). As to transpedicle instrumentation, we modeled instrumentation with a conical body formed by screws of 45 mm in length by 6.9 mm in thickness, with 5 mm bars like beams between both bars that act as transverse connectors (DTT). The screws were placed according to the inward technique, with an internal orientation of about 8 mm. It was assumed that contact between the screws and the bone was perfect, with no mobility. The instruments were made of titanium alloy (Ti6Al4V) with a Young 110 GPa module and a Poisson coefficient of 0.3. This design was chosen since it is easy to copy and is similar to the one used in clinical practice, it is synthetic, intrinsically stable and allows the graft to be tested.

Allograft

As to the allograft, we chose one easy to harvest from the donor, structurally resistant to compression, with a high percentage of cortical bone, of an appropriate measurement and easy to adapt to the interbody space. With all these points in mind, we decided to investigate 4 allograft configurations: the first with a fragment of femoral diaphysis, the second a metapheseal-diaphyseal tibial fragment, the third 3 fibular fragments and the fourth 6 fibular fragments. A 2mm CAT scan was carried out of the femur, the tibia and the fibula of the same patient with the aim of obtaining the internal and external geometrical characteristics, mineral

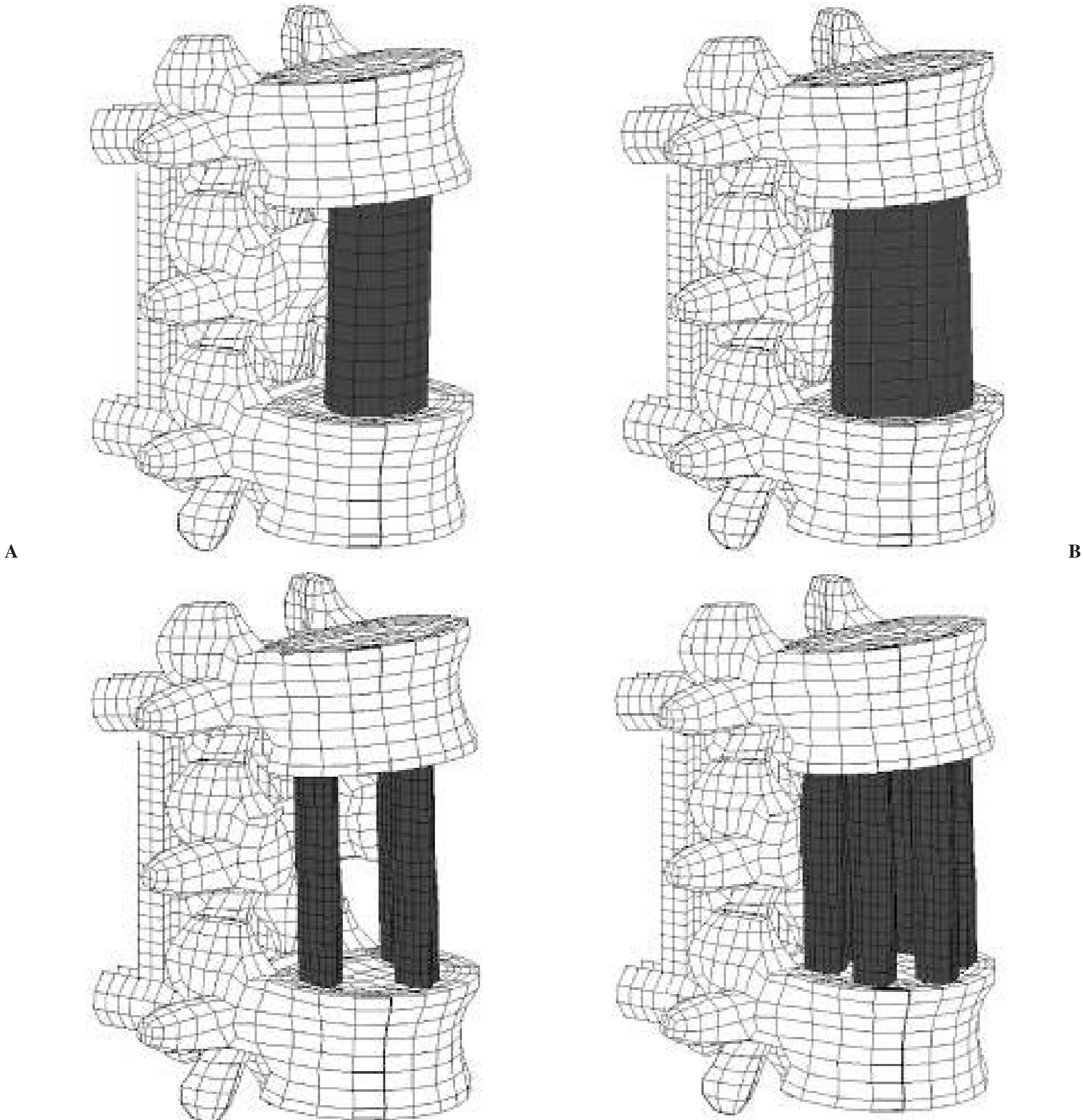

Figure 2. *Finite-element model of the lumbar vertebral segment with allograft (A) femoral shaft, (B) tibial metaphyseal-diaphyseal section, (C) 3 diaphyseal fibular fragments and (D) 6 diaphyseal fibular fragments.*

bone density and the cortical/cancellous ratio of each allograft. Corel Draw was used for the geometry with Bezier type curves and this was exported to DXF type files compatible with AUTOCAD®. The fragments of each allograft were reconstructed and inserted into the model, exported to CDR format with the MENTA© program (MSC Software©) and centered with reference to the geometrical center of each vertebral end-plate (Fig. 2). The bone was considered homogeneous and anisotopic material with the properties defined in Table 1.

Vertebral endplate area

It was assumed that contact between the allograft and both vertebral end-plates was perfect with no mobility, according to the parameters of Chen et al[23], and a solid fusion

Table 1. Properties of the materials used in this study

Material	Composition laws	E (MPa)*	V	G (MPa)
Cancellous bone	Anisotopic elastic (Whyne et al[24], 2001; Ueno and Liu[25], 1987)	140, 140, 250	0.45, 0.31, 0.3	38, 77, 77
Cortical bone	Anisotopic elastic (Ueno and Liu[25], 1987; Natali and Meroi[26], 1993)	8,000, 8,000, 12,000	0.4, 0.35, 0.3	2,000, 2,400, 2,400
Vertebral end-plate	Isotopic elastic (Whyne et al[24], 2001)	1,000	0.3	–
Posterior bony elements	Isotopic elastic (Shirazi-Adl et al[27], 1986)	3,500	0.3	–
CDisc cartilage	Isotopic elastic (Shirazi-Adl[28], 1989; Whyne et al[24], 2001; Natali and Meroi[26], 1993)	24	0.4	–
Facet cartilage	Tension: Isotopic elastic Compression Non-linear elastic	11 (Sharma et al[20], 1995) From 11 to 0% tension from 3,500 to 0.7% tension	0.2 (Li et al[29], 2000) From 0.2 to 0% of tension from 0.4 to 0.7% of tension (Li et al, 2000)	–
Ligaments	Non-linear elastic	Experimental data (Myklebust et al[30], 1988; Chazal et al[31], 1985; Pintar et al[32], 1992)		
Fibers of the annulus fibrosus	Non-linear elastic	Collagen I: Experimental data (Sharma et al[20], 1995)		
Matrix of the annulus fibrosus	Neo–Hookean (Eberlein et al[33], 2001)	μ = 0.5 MPa		
Nucleus pulposus	Mooney-Rivlin (Smit[14], 1996)	C_{10} = 0.12 MPa, C_{01} = 0.03 MPa		
Instruments	Isotopic elastic	110,000	0,3	–
Allograft (Taylor et al[34], 2002)	Anisotopic elastic 11,900 11,900 19,900	0.42 0.230.23	4,000 5,200 5,200	

*The Young module is presented respectively in 11, 22 and 33. The Poisson and Coulomb models are presented respectively in 12, 23, and 31; 1: horizontal direction on the coronal plane; 2: horizontal direction on the sagittal plane; 3: axial direction.

was modeled that would be capable of transmitting loads under compression and tension, thus representing an advanced state of consolidation with the aim of studying the influence of each graft in the long term.

With reference to the treatment of the vertebral end-plate, 4 configurations were assessed: the first one with preservation of the full thickness of the terminal cartilage and supporting the graft on this cartilage; the second eliminating the superficial cartilage layer with preservation of the deep layer, the third eliminating all the cartilage and supporting the graft on the cortical subchondral bone, and, finally, eliminating the subchondral bone and supporting the graft on the cancellous bone of the vertebral body.

Properties of the materials used

All material ratios of force and tension were calculated for long displacements. The properties of the materials based on their composition and the values used to model the different tissues were taken from the literature and are summarized in Table 1[24-34]. The hypoelastic characteristics of the supraspinous, intraspinous, yellow and capsule ligament were calculated based on their biomechanical behavior in flexion for each of these ligaments[35], and from the ratio of the sagittal angle of flexion during mobility of each segment of the spine taking into account the longitudinal stresses of these ligaments[36]. The facet cartilage was defined as hypoelastic under compression.

The absence of linearity was determined by the state of tension according to the direction of contact. Stresses in other directions were calculated my linear elastic isotopic tangential composition laws.

Application of loads and shape changes

Four different types of load were applied, uniformly distributed on the upper end-plate of vertebra L3: a compression of 1,000 N, and movements of flexion, extension and rotation of 15 Nm that represent physiological loads, according to reports in the literature[37]. All degrees of movement of the lower end-plate of vertebra L5 were blocked. Contact similar to that obtained with adhesives was established between the screws and the vertebra (with a distance tolerance of 1.745 mm), contact similar to that obtained with adhesives was established between the allograft and the vertebral end-plate (2 mm) and ordinary contact was established between the joint facets.

All stresses and deformations were assessed by large displacements on each type of graft and type of load using the MSC MARC 2001 program (MSC Software).

RESULTS

Influence of the allograft on the model

First the finite model was tested without adapting either the instruments or the graft, obtaining displacements and stresses for each type of load which were similar to those seen in experimental assays and those described in the literature.

When we introduce into this model rigid posterior instrumentation and an anterior allograft, the total rigidity of the structure is significantly increased. The increase of rigidity on flexion is around 116% with a femoral allograft. With all types of loads and grafts, the stresses undergone by the fixator remain below the failure limit of titanium (795 MPa), with a maximum of Von Mises stresses of 110 MPa with the femoral allograft under flexion loads.

Therefore, using this system no breakage of transpedicle instrumentation is foreseeable, and, given that none of the allografts tested caused failure of the system, this must not be the criteria used for choosing the type of allograft to use. Things are very different when we remove the anterior allograft, since flexion and compression stresses rise above titanium failure limits, about 800 MPa in flexion and about 1,800 MPa in compression. Therefore, the allograft is indispensable to prevent instrument failure under flexion and compression loads.

L4 vertebra displacements were calculated for each type of allograft and load (Fig. 3). The results indicate that flexion causes the load that causes greatest tension on the fixator and the graft. During flexion, the system that uses a femoral allograft has a rigidity of 3,900 Nmm/mm, whereas, comparatively the system using tibia has 15% more rigidity (4,600 Nmm/mm), the system using 3 fibular fragments has 28% more rigidity (5,420 Nmm/mm) and the system using 6 fibular fragments has 33% more rigidity (5,820 Nmm/mm) than the system using a femoral allograft. Under all types of load, the use of 6 fibular fragments has the most rigid configuration, more than the system using femur, or tibia or 3 fibular fragments.

We must add that systems based on instruments and grafts greatly increase physiological rigidity of the spine, elevating it to approximately 10 times more than physiological values (476 Nmm/mm). The variability of the rigidity of the system due to the graft is related to the distance between the centre of the end-plate and the anterior cortical of the graft, the area of end-plate covered and the total thickness of the cortical graft.

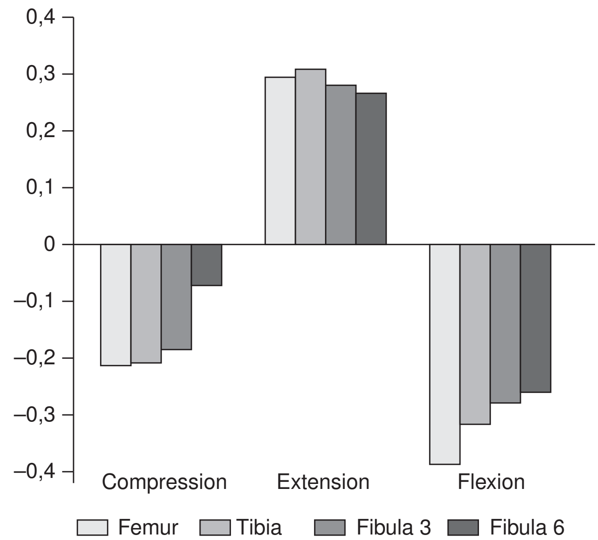

Figure 3. *Axial displacement of vertebra L4 under compression, extension and flexion loads with the 4 types of allograft studied.*

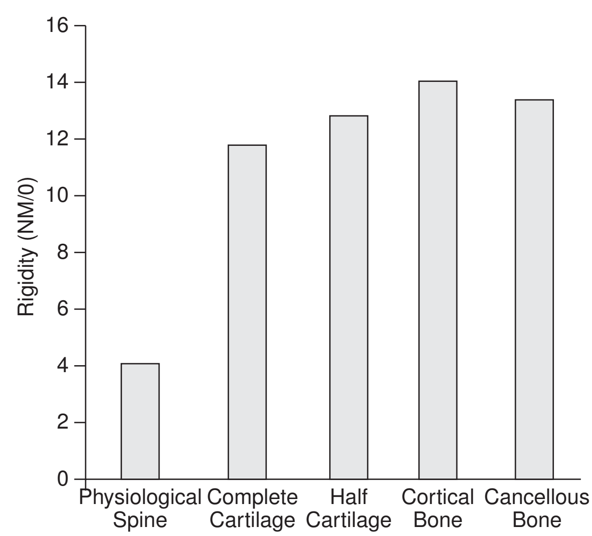

Figure 4. *Rigidity of the segment when end-plate thickness is varied, firstly in a normal spine, and then with femur allograft and a whole vertebral end-plate, with half the cartilage, without cartilage ad without subchondral bone, that is to say using cancellous bone as support.*

Influence of the treatment undergone by the cartilage

If we look at what happens to the variation of thickness of the end cartilage of the vertebral end-plate, we shall see that the rigidity of the system increases as thickness decreases, showing that the end-plate has a load cushioning effect (Fig. 4). The changes in thickness of the vertebral end-plate cartilage result in a change of load transference, with a decrease of load transmission to the posterior instrumentation as the thickness of the cartilage decreases. Von Mises stresses within the depths of the body of L5 (Fig. 5) are homogeneously distributed when we maintain the end-plate cartilage, but, when we support the graft on the corti-

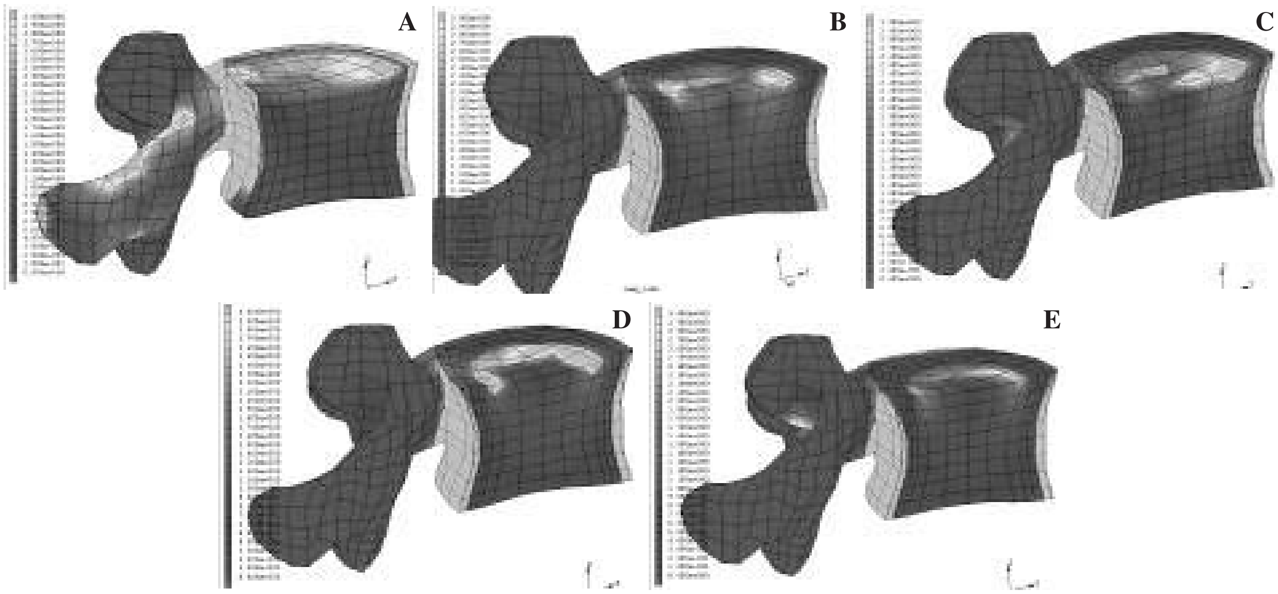

Figure 5. Distribution of Von Mises stresses in the lower vertebra, (A) in physiological conditions, (B) supporting the graft on the whole plate-end cartilage, (C) supporting the graft on half the cartilage, (D) supporting the graft on the subchondral cortical bone, and (E) supporting the graft on subcortical cancellous bone.

cal bone, there are peaks of tension where it comes into contact with the end-plate.

These stresses decrease if we support the graft on cancellous bone, since this has less elasticity and a better distribution of pressure. Therefore, the treatment of the vertebral end-plate that has the best biomechanical result is the preservation of total cartilage thickness (which is not favorable to osteointegration), or elimination until the graft is supported by cancellous bone.

Comparison between grafts

When we compare the distribution of Von Mises stresses in a physiological model *vis a vis* an instrumented model, we see that stresses are decreased within the vertebra and the posterior arc when rigid instrumentation is used.

If we compare the behavior of grafts under tension and compression (Fig. 6) we see that both the femur and the tibia behave in a similar way, but when we test the 3 or 6 fibular segments grafts, we see that the transmission of forces is predominantly through the anterior fibula, whereas the posterior fragments do not bear such a great load, this means that there is greater stress on an area of less graft section and a smaller support area on the end-plate. Due to this there is a greater modification of Von Mises stresses in the adjacent vertebrae.

Sections of L5 show that the distribution of the main stresses with femur and tibia grafts are more similar in distribution to the physiological model than in the systems with fibular fragments, in which behavior is more altered. The same thing is seen with upper vertebra deformities, which follow the same pattern of behavior as the Von Mises stresses (Fig. 7).

In this way, since fibular systems cause higher peaks of tension in adjacent vertebrae, it is to be expected that they will cause a greater alteration of future remodeling patterns that may lead to alterations of the bony arrangement of the spine.

Moreover, Von Mises stresses also increase greatly if we change the orientation of the anterior diameter of asymmetric grafts. When we rotate the tibia 90°, Von Mises stresses within the upper vertebrae increase; which indicates that the position of the graft is an important factor in load transfer (Fig. 8). In this way, the use of tibia implies a greater asymmetry in displacement and Von Mises stresses, either in cases of tension or compression. Therefore, femoral grafts limit this imbalance when compared to tibial grafts since they are more cylindrical. In the case of tibial grafts, the asymmetric geometry of the graft predisposes to rotational instability of the forces due to its shape (Fig. 9).

DISCUSSION

A finite elements analysis has been performed of anterior vertebral fusion with substitution of a spondylectomy by different grafts. It is evident, from this study, that the use of allografts combined with posterior instrumentation drastically changes the transfer of loads under physiological conditions. Furthermore, the use of allografts with a different

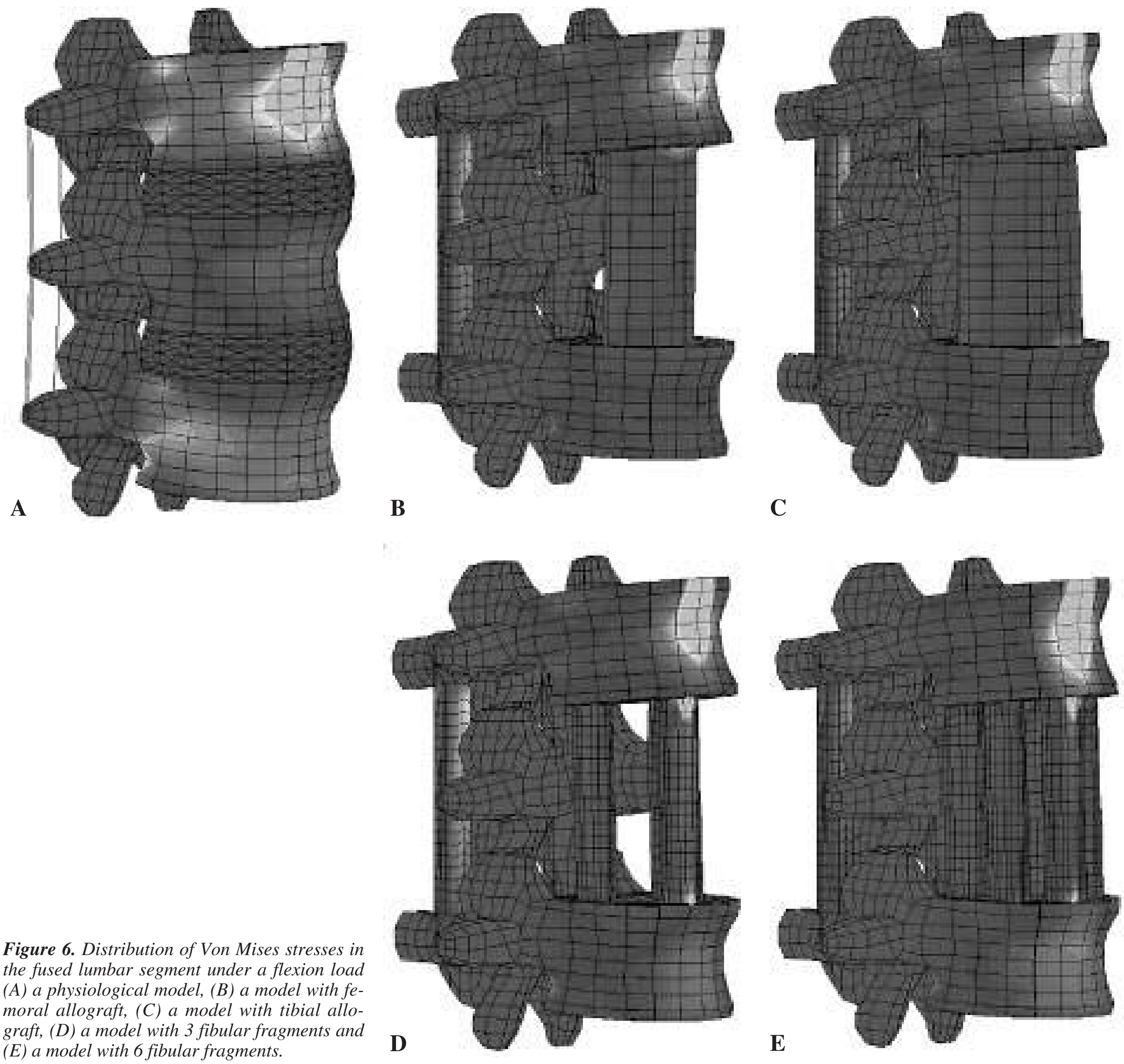

Figure 6. *Distribution of Von Mises stresses in the fused lumbar segment under a flexion load (A) a physiological model, (B) a model with femoral allograft, (C) a model with tibial allograft, (D) a model with 3 fibular fragments and (E) a model with 6 fibular fragments.*

geometry and position may lead to important differences in spine biomechanics.

Many studies have compared the behavior of anterior allografts in the lumbar spine, but most of them have done it from a biological or clinical point of view[3-6,38]. We have only found 2 studies that analyzed allografts comparatively from the biomechanical point of view, both in experimental models. Rao et al[39] compared femoral allografts with one fragment fibular allografts and iliac crest allografts and concluded that the femur is the most appropriate, since it presents greater rigidity and fails at higher loads, in spite of the fact that its resistance (load failure standardized by area) is similar.

In our study we saw evidence that fibular grafts provide the greatest rigidity, either of 3 or 6 fragments. This correlates with the work carried out by Siff et al[40], who compared a femur diaphyseal fragment with 6 fibular fragments in the substitution of the anterior part of the vertebral body. These authors concluded that fibular fragments are biomechanically superior since they form a more rigid and stable system, in spite of the fact that this is not statistically significant; therefore, these authors support the use of fibular fragments instead of femoral allografts.

Our results coincide as far as the increase of rigidity with fibular fragments, but we do not coincide with the conclusions of this study since the increase in rigidity causes an increase in tension and deformation of adjacent vertebrae and is therefore a negative factor in the long term. Therefore, in the long term, the use of fibular fragments can con-

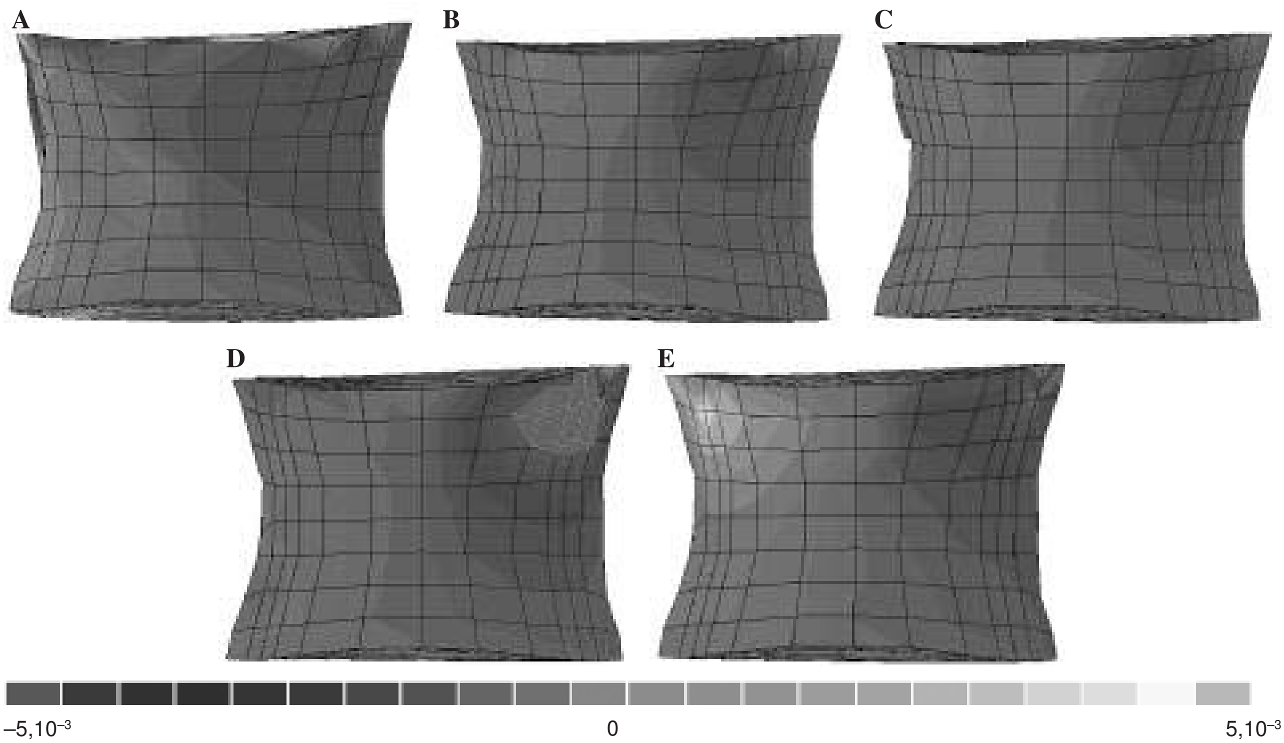

Figure 7. Distribution of deformations in the vertebral body of L5 under a compression load (A) a physiological model, (B) a model with femoral allograft, (C) a model with tibial allograft, (D) a model with 3 fibular fragments and (E) a model with 6 fibular fragments.

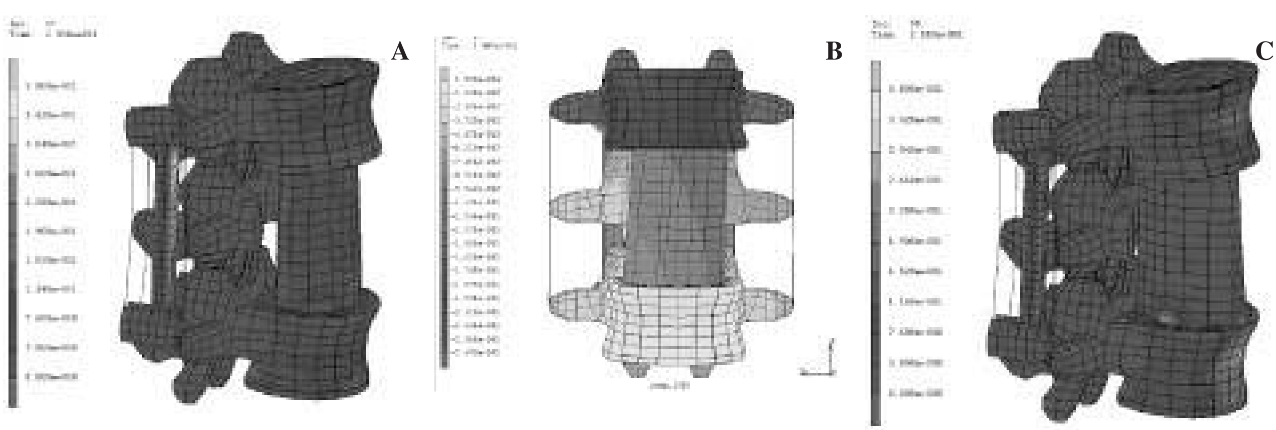

Figure 8. Distribution of Von Mises stresses in the tibia model under a compression load, (A) lateral view with the tibia placed with its major axis coronally, (B) frontal view with the tibia placed with its major axis coronally, in which tension asymmetry may be seen and (C) lateral view with the major axis placed sagittaly.

dition remodeling of the upper vertebrae as an adaptive mechanism to tension alteration.

On the other hand, femur and tibial grafts are better for preserving tension and deformation distribution in adjacent vertebrae. However, we have seen in this study that asymmetric tibial geometry may include tension and biomechanical imbalance, and this can worsen if the surgeon, during the operation, places the allograft in an irregular manner. Asymmetry can be detrimental to the load transfer process and create instability in the long term.

As to the position of the graft, Cunningham and Polly[41] studied the effects, from the mechanical point of view, of placing titanium cages in the posterior, mid and anterior interbody space. These authors saw that rigidity during axial compression was 750 kN/M when the cage was placed posteriorly, it increased to 1,600 kN/M in the mid zone and in-

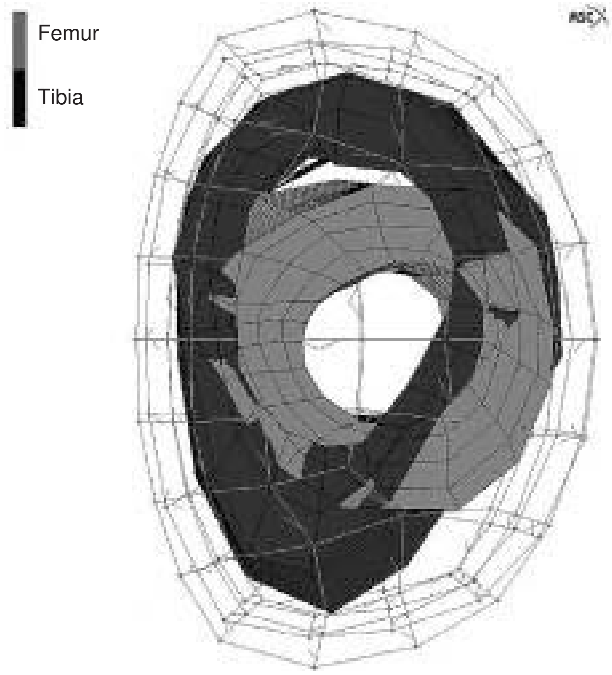

Figure 9. Comparison of the geometry between a femur and tibia allograft with reference to the position of the vertebral end-plates and the anterior zone of the body.

creased up to 2.000 kN/M in the anterior zone. The results indicate that the rigidity of the assembly is highly sensitive to the position of the cage in the interbody space in the sagittal plane and increases linearly as we place the graft further forwards. These results completely coincide with what we found in our study. Another contribution that substantiates this is the work of Zander et al[8,42], who used a finite element model to analyze 4 conditions of interbody grafts. These authors found that tension distribution is similar except when the graft is placed in the mid sagittal zone, where stresses are very much higher. Therefore this study confirms our results as to the fact that the position of the graft has an important effect with reference to the distribution of tension in adjacent vertebrae and when desiring to minimize mobility in the segment on which we wish to carry out arthrodesis.

The thickness of the cartilage of the end-plates plays an important role in load transfer as a shock absorber in the structure. From the biological point of view it is advisable to remove the maximum amount of cartilage to stimulate consolidation. From the biomechanical point of view it is advisable to support the graft on cancellous bone rather than cortical bone, since in this manner the transmission of loads and stresses in the adjacent vertebrae is not altered so much and the rigidity of posterior instrumentation will prevent the collapse of the assembly. This factor will also favor consolidation due to the greater amount of vascular structures in cancellous bone.

As to treatment of the vertebral end-plate, Oxland et al[43] biomechanically analyzed its influence on the properties of vertebrae. The results showed that, when the vertebral end-plate is removed, resistance to vertebral failure becomes reduced, but this reduction is only significant in the posterior part of the vertebrae, not in their center or anterior zone.

Rigidity or stiffness is also significantly reduced, with an inverse distribution to resistance. The authors concluded that the removal of the vertebral end-plate has a negative effect on structural resistance of the vertebra and increases the risk of subsidence of the graft, especially in the posterior zone, although not so much in the central zone. One point to criticize in this study is that it was carried out in elderly patients (72 and 82 years), with low bone mineral density of the underlying vertebral body and in a non-instrumented model.

On the other hand, Hollowell et al[44], in an experimental cadaver model, studied the contribution of the vertebral end-plate with grafts and anterior cages in the thoracic spine in 63 vertebrae (in patients with a mean age of 63 years). The authors conclude that preservation of the vertebral end-plate is not an important contribution to the reconstruction of the anterior spine nor does it prevent graft subsidence. Finally, Polikeit et al[45], using a finite elements model, investigated the properties of the vertebral end-plate and tension distribution in the lumbar spine with an interbody implant. The authors concluded that the presence of a rigid vertebral end-plate tends to increase stress concentration and to create high pressure zones.

Since this is a finite elements study, we wish to point out some of its limitations. The physiological three-dimensional model that was modified to be able to insert transpedicle instrumentation was based on a single individual. This model has material properties, contact and geometry characteristics that may vary from one patient to another. However, this model represents the ideal lumbar spine modeled by finite elements and the properties of the materials were taken from the literature with the aim of representing the best average of a normal population. Transpedicle instrumentation was defined as completely bone anchored, with no pre-tension. Although pre-tension may modify the results of stresses within the graft[42], in this study, which is mainly comparative, the results obtained comparing several allografts do not vary. This is especially true given that the contact conditions between the graft and the vertebral end-plate were assumed to be perfect, creating a model with effective fusion. A finite elements analysis allows us to eliminate possible errors from the experimental study such as irregular sections or asymmetric grafts, the difference in longitude between several fibular fragments, the variations in the placement of instrumentation between models and variations in simulation of consolidation in the long term. Moreover, it contributes a very superior set of data to those obtained from an experimental study with relation to what really happens within adjacent vertebral bodies.

The analysis of stresses in this study provides a quantitative indicator of stresses in the fixator. It has been seen that stresses within the fixator remain below the resistance and fatigue limit in all cases. Therefore, not only must transpedicle instrumentation be designed with the purpose of reducing interior stresses, but also keeping in mind the influence the fixator has on other spine components. In all types of load, the use of a fixator combined with allograft, considerably changes stresses and deformations, which in the long term may involve an adaptive process that can be simulated using a remodeling algorithm[46,47].

CLINICAL IMPLICATIONS

This study has some clinical implications. During the last few years the number of circumferential fusions performed has increased, many of them have used anterior structural allografts[48]. This study corroborates the concept that femoral graft, or any cylindrical section of graft in its place, is the most favorable graft, from the biomechanical point of view, to achieve circumferential fusion. These biomechanical results are supported by results seen during clinical practice, in which we have found that femoral grafts have an excellent outcome during long term followup[38]. Therefore, femoral grafts should currently be the first choice for anterior substitution of the vertebral body, in comparison with other grafts, such as fibular grafts, which must only be used when there are problems with adaptation to the interbody space with femoral grafts and it is necessary to use small supports with a smaller diameter.

Another important issue is treatment of the vertebral end-plate. According to our results, resection of the vertebral end-plate, when we put in place rigid titanium instrumentation posteriorly, does not cause any problem as far as the stability of the assembly. Therefore, in clinical practice, if we completely resect the vertebral end-plate and support the allograft on cancellous bone, this will have 2 advantages over the preservation of the vertebral end-plate: a) a greater biological advantage as far as obtaining a vascularized bed to achieve integration of said graft, and b) greater biomechanical advantages since there is a better stress distribution on the vertebral body due to the cancellous bone having less elasticity. Therefore, this may reduce the rate of malunions of anterior grafts in spine surgery, since it will presumably favor consolidation. This is, therefore, the second factor that we think important in clinical practice, since many authors prefer to preserve the vertebral end-plate[3,4,6], with the consequent risk to graft consolidation. However, in contrast with what has been seen with the use of femoral grafts, up to date, these experimental results obtained *in vitro* with the resection of the vertebral end-plate have not been corroborated by clinical studies. In conclusion, the use of analysis by finite elements is a useful clinical tool for biomechanical studies for choosing one allograft over another. This type of study has the advantage of only being able to change a few controlled parameters and to rule out many other variables which are difficult to control in experimental or clinical studies, such as characteristics of the donor (rigidity, measurement, age, bone mineral density, etc.), graft treatment conditions and issues related to the placement within the interbody space (contact between graft and end-plate, location, load asymmetry, etc.). We may therefore conclude that femoral diaphysis fragments used to substitute the vertebral body modify the stress distribution in adjacent vertebrae to a lesser degree, in comparison with tibial and fibular grafts. We may also conclude that the preservation of cortical bone of the vertebral end-plate possesses no significant biomechanical advantage in anterior spine reconstruction when rigid transpedicle instrumentation is used.

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