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The inverse effects of load transfer and load sharing on axial compressive stiffness

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Abstract

Background context: The goal of spinal instrumentation is to stabilize involved motion segments while fusion occurs. Although some degree of load sharing is necessary for fusion, the ability of the instrumentation system to transfer the load may vary.

Purpose: The purpose of this study is to formulate a mathematical relationship between load sharing and load transfer of specific spinal instrumentation systems using a well-accepted mechanical model.

Study design/setting: Forty-eight American Society for Testing Materials standard ultra high molecular weight polyethylene cylinders were used as per designation F 1717-96, standard test methods for static and fatigue for spinal implant constructs in a corpectomy model.

Methods: Twenty-four spinal assemblies consisting of anterior plates, anterior rod, and posterior rods were subjected to compression bending tests using a MTS Bionix servo-hydraulic material testing apparatus. Each implant was tested in compression bending with and without the addition of a titanium load-sharing cage. The force applied was the independent variable, and the displacement was the dependent variable. The stiffness was determined for each setup with and without the addition of an anterior load-sharing cage.

Results: The average axial compressive stiffness of a system increased by a factor of 8.5 with the addition of the load-sharing cage. An inverse relationship existed between the compressive stiffness of the construct and its relative increase achieved with the addition of the load-sharing cage. The compressive stiffness of the system with the addition of the load-sharing cage approached that of the anterior device itself as the system flexibility increased. The ability of instrumentation systems to load share or load transfer and their respective stiffness was determined.

Conclusions: The 5-mm rod screw posterior system was compared with the 7-mm Ti posterior system with the addition of one and two devices for transverse traction (DTTs). The rods with the increased diameter had a stiffness of 1723 n/mm with one DTT and 1815 n/mm with two DTTs. The addition of an anterior cage had little effect on the stiffness of these systems. Anterior plate and screw/rod systems were analyzed and showed similar mechanical behavior to the 5-mm posterior rod/screw systems. A significant increase in stiffness was realized with the addition of an anterior cage. A means to determine the load sharing/transferring properties of a spinal instrumentation system is presented. This technique will allow the amount of load transferred from the fusion mass to the instrumentation to be predicted. © 2001 Elsevier Science Inc. All rights reserved.

Keywords: Instrumentation; Load transfer; Stress shielding; Load sharing; Axial compression; Stiffness

Introduction

Clinical instability is defined as the loss of ability of the spine under physiological loads to maintain its pattern of displacement so that there is no initial or additional neurological deficit, no major deformity, and no incapacitating pain [1,2]. The goal of spinal fusion is to restore stability to

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a clinically unstable spine, with a minimum decrease in spinal motion and minimal disruption of normal structure and function of the spinal column [2]. Spinal fusion maintains correction, provides structural integrity, and by diminishing the spine's movement may alleviate pain [3].

Spinal fusion is achieved with much greater certainty by the use of internal fixation [4–8]. The stiffness of a spinal fusion construct (spinal column with instrumentation) is perhaps the most significant factor in success of spinal fusion [2]. Stiffness of the construct is determined by the design of the device as well as the anatomic considerations, such as location and quality of existing bone and graft [2]. Securely applied spinal instrumentation and the fixated bone behave like a composite. The bone would be expected to share the load with the implant in proportion to the ratio of their stiffness [7]. Stiffness is mathematically defined as the ratio of resistance (to an applied load) offered to the displacement imposed [9].

Higher implant stiffness is associated with a higher fusion rate, but lower density of the underlying bone. A less rigid system has been correlated to increased bone density [3,10,11]. Somewhere in between these two extremes would be an ideal construct for successful fusion, applicable to the individual patient medical situation. Therefore, the purpose of this study is to use a well-accepted comparative mechanical model to create a mathematical relationship between stress shielding and load sharing in the spine. The model can then define a spectrum of load-sharing and stress-shielding properties of spinal constructs.

Materials

F 1717-96 standard test methods for static and fatigue for spinal implant constructs in a corpectomy model were used to perform a compression bending test on three groups of thoracolumbar and lumbar spinal implants [12]. Each pair of superior and inferior American Society for Testing Materials (ASTM) standard 62 mm diameter × 35 mm depth ultra high molecular weight polyethylene (UHMWPE) cylinders was instrumented with an anterior plate, anterior rod construct, or a posterior rod construct. This produced one above/one below corpectomy models. Figs. 1 demonstrate two examples of the corpectomy models instrumented with three different types of instrumentation with and without the titanium Harms cage. The blocks were attached to the jig using a ball-socket joint on the actuator. Axial loads (compressive bending as per ASTM guidelines) were applied. The test machine applies loads in the Z direction without constraining rotation in the X-Y plane. The jig also allows motion in the X-Z plane as per ASTM guidelines [12].

The standard test methods for static and fatigue for spinal implant constructs in a corpectomy model set out guidelines for load types and methods of applying loads. Methods for three static load types and one fatigue test are defined for the comparative evaluation of spinal implant assemblies. The standard used in this study and recommended by the ASTM establishes guidelines for measuring displacements, determining yield load, and evaluating the stiffness and strength of spinal implant assemblies. The three static mechanical tests described in section 4.3 of the guidelines include compression, tensile, and torsion. The authors chose compression testing to determine the mechanical parameter of stiffness in the spinal implant assemblies. The results of the compression testing can be used to compare different component designs in terms of the relative mechanical parameters, in this case stiffness (section 5.3) [12].

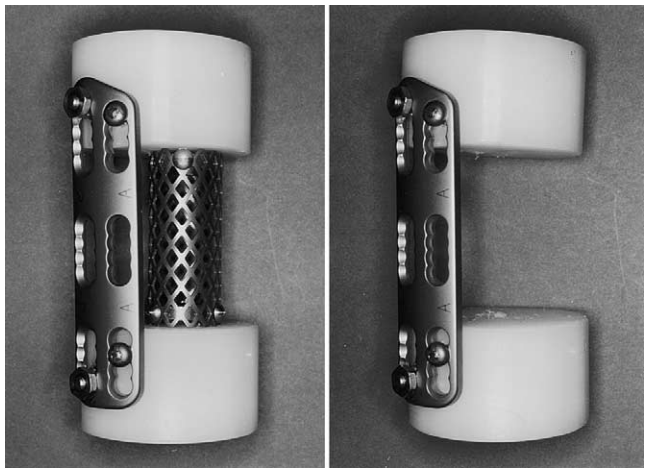

Fig. 1. Anterior device with and without a Harms cage.

Each implant was tested with and without the addition of a titanium Harms (Depuy, Warsaw, Indiana) load-sharing cage. The vertical spacing between the blocks was 50 mm as determined by the implant. The implant (anterior cage) was placed consistently to establish a constant block moment arm from the cage to the anchors of the implant. The posterior aspect of the cage was 5 mm from the posterior midline border of the UHMWPE block. This placed the cage directly beneath the actuator.

The gap between the two simulant blocks was 5.0 cm: 3.0 cm for the missing vertebra and 2 cm for the two missing discs. The systems used were (a) an anterior Z-plate, (b) an anterior university plate, (c) an anterior synthes plate, (d) an

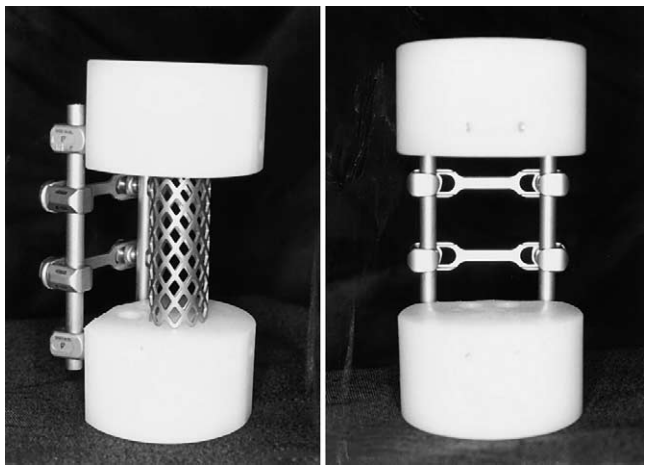

Fig. 2. Posterior device with and without a Harms cage.

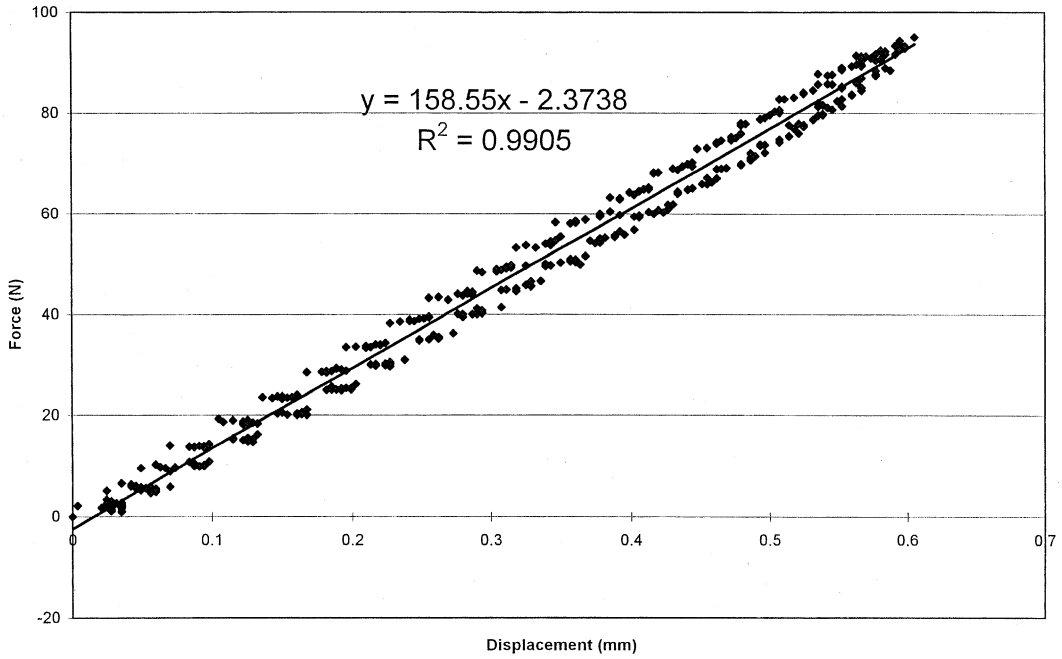

Fig. 3. Load–displacement graph for a synthes plate without a Harms cage.

anterior 5-mm Moss Miami rod with uniaxial screws, (e) a posterior 5-mm Moss Miami with polyaxial screws, and (f) a posterior universal 7-mm rod with one cross connector, and a posterior universal 7-mm rod with two cross connectors. The six spinal assemblies were tested as per the ASTM guidelines. Four spinal assemblies were tested in each of the above six subgroups. All spinal assemblies were assembled by a single operator according to manufacturers' specifications. All components in the spinal assembly were previously unused and were not retested. All UHMWPE test blocks were used for one test only.

All mechanical testing was conducted using an MTS machine 858 Mini-Bionix (MTS Systems, Corp., Eden Prairie, MN) bi-axial, servo-hydraulic material testing apparatus. A ramp of 1 N/s was applied until an axial compression preload of 10 N had been established. One preconditioning was run followed by 10 cycles between 10 N and 110 N at a frequency of .25 Hertz. Examination of the load-deflection curve may reveal laxity in the fixture. The linear portion of the curve will define the straight-line section of the load-displacement curve after the laxity has been removed. The preconditioning cycle establishes if laxity is present.

The compression tests recorded force (N), which was the independent variable, displacement (mm), which was the dependent variable, and time (s), by means of MTS TestStar SX II version 3.0C. Data was collected at a frequency of .25 Hertz using TestWare SX for OS/2 operating system version 3.0C(MTS Systems, Corp., Eden Prairie, MN).

The data were directly downloaded into and was then statistically analyzed using Microsoft Excel for Windows version 7.0. The average stiffness was determined for each setup. Axial stiffness was defined as the gradient of the force (N)-versus-displacement (mm) curve.

The use of UHMWPE cylinders instead of cadaveric spines provided consistent fixation, eliminating biological deterioration during testing and specimen variables of bone mineral density, morphology, and age [13]. By normalizing the specimen, the protocol was designed to test the mechanical properties of the constructs only, but not the bone.

Results

The stiffness of each system was plotted against the change in stiffness with the addition of an anterior cage. Figs. 4 and 5 show the load-displacement graphs of two different systems. If the construct showed little or no change in stiffness with the addition of an anterior cage, then the construct was load transferring.

If the addition of the cage increased the stiffness of the system, then the load was being shared by the system and the cage. Load sharing would exist in this system.

The average stiffness for each system is shown in Table 1. The 5-mm posterior rod system with polyaxial screws had the lowest stiffness of all systems tested with a stiffness of 59 n/mm. With the addition of an anterior cage, the stiffness of the system was increased by a factor of 10.88 to a stiffness of 642 n/mm. The 5-mm rod screw system was compared with the 7-mm Ti posterior system with the addition of one and two DTTs. The rods with the increased diameter had a stiffness of 1723 n/mm with one DTT and 1815 n/mm with two DTTs. The addition of an anterior cage had little effect on the stiffness of these systems.

Anterior plate and rod screw systems were analyzed and showed similar mechanical behavior to the 5-mm posterior rod screw systems. A significant increase in stiffness was realized with the addition of an anterior cage.

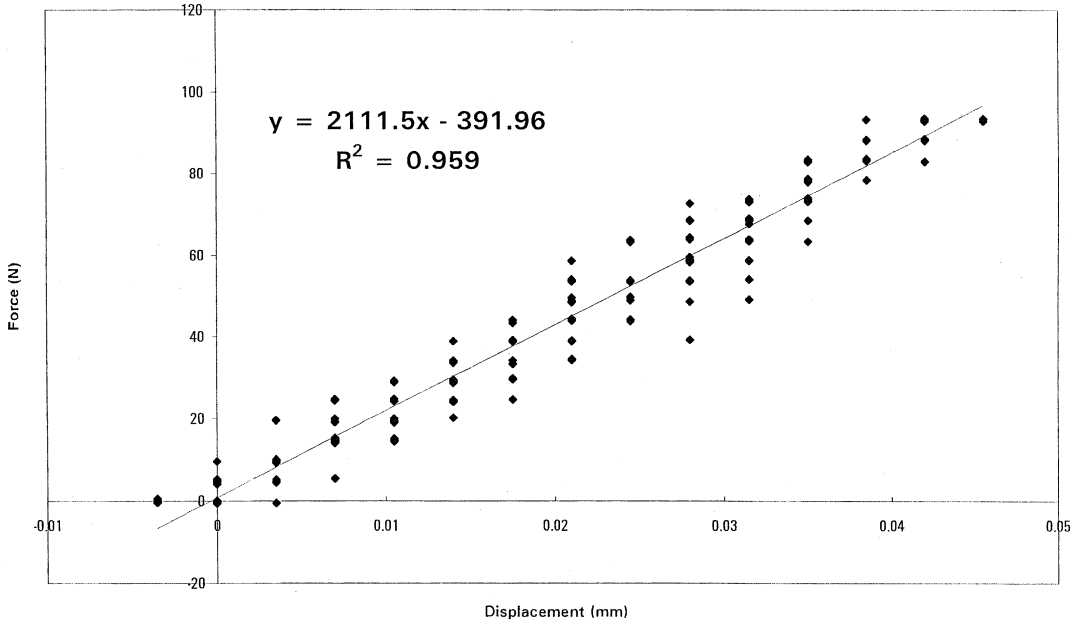

Fig. 4. Load–displacement graph for a titanium with two cross connectors with a Harms cage.

The device was defined as load sharing when a significant increase in stiffness was present with the addition of an anterior cage. On average, the axial compressive stiffness of the load-sharing system was approximately 8.5 times greater with the addition of the load-sharing device. However, when the compressive stiffness of the implant construct was 2000 N/mm or above, the addition of a Harms cage had no effect on the compressive stiffness of the system. The system was considered load transferring.

The stiffness of each system was plotted against the change in stiffness with the addition of an anterior cage. The inverse relationship between the axial compressive stiffness of the im-

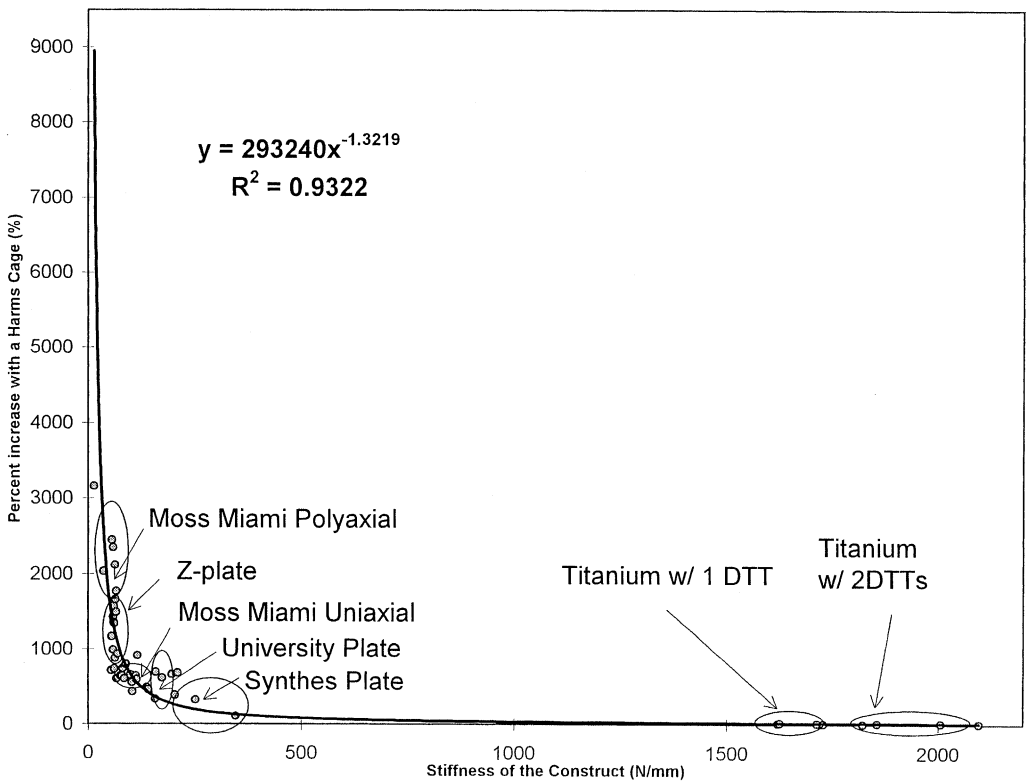

Fig. 5. The inverse relationship between the stiffness of the system and the percent increase in stiffness with the addition of a Harms cage.

Table 1
Average stiffnesses (N/mm) and significance between the constructs with and without a Harms cage

	Without Harms cage	With Harms cage	Comments	*p* value
Moss Miami polyaxial	59	642	Posterior	.004
Z-plate	63.5	869	Anterior	.01
Moss Miami uniaxial	103	675	Anterior	.03
University plate	141	791	Anterior	.008
Synthes plate	158	685	Anterior	.001
Titanium with one cross connector	1723	1955.5	Posterior	.07
Titanium with two cross connectors	1815.33	2061.5	Posterior	.08

plant construct and the relative increase in axial compressive stiffness achieved with the addition of the load-sharing device is graphically presented in Fig. 3. The mathematical equation for the inverse relationship was found to be:

(% increase in the stiffness of the whole system with the addition of a load-sharing device) $\times$ (stiffness of the system)$^{1.32} = 293{,}240$

This mathematical model had an R^2 of .93.

Discussion

The ability of bone to form optimal structures to support loads and to adapt structurally to changing loads has been qualitatively described by Wolff's law [14]. Wolff's law of trabecular architecture states that bone adapts its structure to mechanical loading conditions; axial load sharing is necessary for a fusion to occur. In normal bone, a continuous remodeling process, known as the strain-adaptive remodeling theory [11,14], is occurring resulting from normal physiologic osteoclastic and osteoblastic activities.

The UHMWPE test blocks eliminate the effects of the variability of bone properties and morphometry. The standard is used to quantify the static and dynamic mechanical characteristics of different designs of spinal implant assemblies. If the moment arm from the point of load application to the anchor is kept constant, then test data from different spinal assemblies with different intended methods of applications may be compared. Then results may also be used to compare different component designs in terms of relative mechanical properties, in this case the changes in stiffness of the spinal assembly with and without anterior support. A securely fixed load-sharing implant and bone behave like a composite. The bone would be expected to share the load with the implant in proportion to the ratio of their stiffness.

It must be stressed, however, that mechanical tests conducted in vitro use simplified load schemes and do not attempt to mimic the complex loads of the spine. The tests in vivo will differ from the loading configurations used in the ASTM standard. The results cannot be used directly to predict in vivo performance. The results can be used to compare different component designs in terms of relative mechanical parameters.

The authors have shown that an inverse relationship between the stiffness of the implant and the increased load-sharing effect on the system has been investigated and established in the present study as follows:

(% increase in the stiffness of the whole system with the addition of a load-sharing device) $\times$ (stiffness of the system)$^{1.32} = 293{,}240$

The graph representing this relationship shows the stiffness for each particular system with and without an anterior cage. The construct or assemblies found on the lower portion of the graph (small *y* value) did not have a significant percent change with the addition of a anterior cage. These assemblies were considered load transfer, because no load was transferred to the cage as evidenced by no change in stiffness with the addition of a cage. The assemblies on the upper portion of the curve (large *y* value) did reflect a change in stiffness with the addition of an anterior cage. These would be considered load-sharing systems, because a significant portion of the load was being transferred to the cage. The mathematical equation is an exponential regression or a decaying exponential. The slope of the graph is 293,240. There is therefore a relationship between the *y* and *x* axis. This establishes an inverse relationship between the percent increase in stiffness with an anterior cage and the original stiffness of the system. It is not a formula but it represents and more importantly establishes the relationship between load-sharing and load-transfer systems.

As axial stiffness of the implant increases relative to the axial stiffness of the vertebral column, a smaller percentage of the axial load will be transmitted through the vertebral column [5]. This trade-off relationship between load sharing and stress shielding creates a spectrum of construct combinations for surgeons to select or use as a reference before surgery.

The optimal stiffness level remains to be established in terms of development of bony union, the prevention of excessive stress shielding, related osteoporosis, intervertebral deflections, and the implant failure [2]. The possible contribution of the implant stiffness toward increased rate of degeneration of disks or facets at the ends of the fusion is also crucial.

It has been hypothesized that the rigidity of the fusion device should vary throughout the different phases of healing [15]. Because the fusion mass initially has no strength, the load is mainly shared by the implant and the spinal column. Effective fusion, therefore, requires early control of motion. A structure that is more rigid at the beginning of healing will provide early stabilization and encourage the formation of delicate vessels. However, in later stages, the fusion mass gains stiffness and takes over more of the load with time, thus unloading the implant. The construct should then allow transfer of loads through bone. A fixation device should be able to provide a relatively stable environment at the fracture site that will enhance healing potential and result in bony healing before the device fails [16]. Load-transfer devices may be indicated when the spine is unable to support physiological loads as a result of trauma or malig-

nant destruction of all columns of the spine. Load-sharing devices may be indicated when bony continuity exists or the anterior column has been re-established with bone grafts.

In the case of a rigid construct, the biological requirement for the formation of callus to increase the rigidity of the fracture site is reduced, resulting in a mobile and unstable bone–construct healing interface [17].

However, according to Duffield et al. [9], an implant can be "too stiff" in terms of maximizing the axial stiffness of the fused spine that is achievable in 24 weeks. They observed that initial bone mineral content loss up to 12 weeks, with implant removal at 12 weeks producing a statistically significant rebound, while leaving the implant in place resulted in no further bone loss [9].

According to the studies assuming concentric cylindrical geometry for the bone and prosthesis [1], the stress-shielding effect can be biomechanically reduced by using a lower modulus device if possible. To reduce the extent of stress shielding in a more flexible and clinical manner, interbody bone graft alone or in combinations with a fixation device could be applied to transfer stresses at the adjacent levels [13,18], which in turn spreads out the stress concentrated at the screw–bone interfaces.

The selection of an implant creating the desired stiffness ratio between the implant and the host bone depends on the trade-off between potential effects of stress shielding relative to the fatigue strength/life of the selected implant [13,16]. If the magnitude and frequency of the spinal loads exceed the fatigue tolerance limit of the implant, the implant will fail. The selection of an implant also depends on the type of activities in which the patient is involved and number of cycles that the implants needs to tolerate until fusion has satisfactorily been achieved [9].

If a fusion is to be obtained in the case of a degenerative spine, the role of the instrumentation is to prevent translation of the motion segments, thereby allowing the compressive loads to be transmitted through the vertebral bodies and the degenerated disks [3]. Therefore, less stiff spinal constructs are preferred. In the case of a fracture, the instrumentation is made to withstand the bulk of the loads, because the structural integrity of a motion segment may have been lost. Stiffer constructs are preferred to support and to protect the fractured motion segment [3,19].

Conclusions

It has been seen that higher implant stiffness is associated with higher fusion rate but results in a lower bone density. A less rigid system has resulted in increased bone density [3,10,20]. A mechanical model was created in the present study to illustrate the mathematical relationship between stress shielding and load sharing in the spine, which provides an understanding of the spectrum of instrumentation combinations. Surgeons can then select the appropriately stiff construct for each individual patient situation.

The maximum effect on load sharing and load transfer was determined for anterior and posterior spinal constructs. The compressive stiffness transmitted through the anterior spinal column was calculated using cages. The difference between the construct stiffness with and without the anterior cage quantified the load-sharing and load-transferring properties of the construct.

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