Andrew J. Rapoff
Wesley M. Johnson
Jeremy Handel
Raymund Woo

Interbody allograft in a skeletally immature spine model

Received: 26 February 2002
Revised: 16 September 2002
Accepted: 27 November 2002
Published online: 2 April 2003
© Springer-Verlag 2003

A. J. Rapoff (✉) · W. M. Johnson
Department of Mechanical
and Aerospace Engineering,
Department of Biomedical Engineering,
University of Florida, Florida, USA
Tel.: +1-352-3920961,
Fax: +1-352-3927303,
e-mail: rapoff@ufl.edu

A. J. Rapoff
University of Florida,
231 MAE-A Box 116250,
FL 32611-6250 Gainesville, USA

J. Handel
College of Medicine,
University of Florida, Florida, USA

R. Woo
Department of Orthopaedics
and Rehabilitation, University of Florida,
Florida, USA

Abstract The objective of this cadaveric biomechanical study was to establish further bovine spines as models for evaluating lumbar interbody allografts and to provide guidance for their use in pediatric humans. It is unknown whether interbody allografts can be used in the pediatric spine without failure of the host vertebral bone. Allografts were placed in cow and calf spines and loaded in compression. The cow spines were much stronger and stiffer than the calf, but moderate in vivo activities were estimated to result in loads on the allograft constructs that would result in host bone failure. Bovine spines were established as suitable models for the compressive behavior of interbody allografts in the human spine, when bone density is considered. Interbody allografts should continue to be used with ad-

junctive instrumentation so as to preclude host bone failure.

Keywords Biomechanics · Lumbar spine · Interbody allograft · Compression failure

Introduction

Interbody cortical bone allograft use in lumbar spine procedures has grown tremendously in the past decade. Although not delineated by musculoskeletal graft type, the American Association of Tissue Banks reports a 267% increase in the number of musculoskeletal allografts distributed nationwide from 1994 to 2000 (Personal communication, American Association of Tissue Banks, 2001). Two organizations state that they ship between 90,000 and 200,000 processed allografts each year [17, 19]. The reasons for this increase are many, and probably include the enhanced awareness of organ donation through driver li-

cense programs, the ability to harvest allograft bone in much greater quantities than autograft bone [6], the ability to precision machine the allografts to consistent dimensions, the increase in number of tissue banks and processing organizations, and the marketing efforts of the allograft organizations themselves.

Despite this increased use, interbody allograft bone is known to incorporate more slowly, incorporate to a lesser histologic degree, and possess decreased radiographic fusion rates compared to autograft bone [3, 6]. These realities are especially evident in posterior lumbar spine procedures in adults, but manifest themselves also in anterior procedures in which adjunctive instrumentation is not applied. On the other hand, corrections of thoracolumbar de-

formities in pediatric patients involving interbody allografts coupled with posterior instrumentation has proven to be a successful procedure with respect to maintaining correction and incorporation of the allograft [1, 3, 10], in part due to the increased bone remodeling frequency in the pediatric population [7].

Load sharing offered by adjunctive instrumentation may play a role in decreasing deleterious stresses at the allograft-host vertebral body interfaces, especially if acute postoperative loading is such that failure of the host bone occurs. The lower body weight and lesser muscle mass of the pediatric anatomy means less severe spinal loading compared to the adult spine. This lower loading may allow for the reduced size implant called for in the smaller pediatric spine, whose host bone strength also is reduced due to incomplete mineralization and lower apparent density, which combine to produce lesser strength. Thus, it is unclear as to whether interbody allografts are appropriate as stand-alone implants in the pediatric spine.

In fact, the use of interbody cortical bone allograft has not been investigated in skeletally immature spines in a controlled *ex vivo* laboratory setting. One issue that must be addressed prior to their use without adjunctive instrumentation is the ability of the pediatric host vertebral bodies to withstand compressive loads associated with activities of daily living. Therefore, the first purpose of this project was to compare the compressive load bearing properties of an interbody allograft when placed either anteriorly or posteriorly in immature (calf) or adult (cow) bovine cadaveric lumbar spine units. The mechanical behaviors of the immature spine allograft constructs were hypothesized to be similar to those of the adult spines. The second, yet primary, purpose of this project was to attempt to establish the use of bovine spines as comparable models for further studies of interbody allografts in humans. The motivation for this is the difficulty and expense associated with obtaining cadaveric human spines, especially pediatric specimens. To this end, a compressive strength-apparent density regression relation was constructed from the bovine data. Previously published values for the apparent density of human vertebral trabecular bone were entered into the developed regression relation to estimate their strength. These strengths were compared to compressive stresses expected to act on human vertebrae.

Materials and methods

Allograft description

The allograft selected for use in this project is the Tangent impacted cortical wedge (Fig. 1; Regeneration Technologies Incorporated; Alachua, Fla.; represented by Medtronic Sofamor Danek (MSD); Memphis, Tenn.). The Tangent can be ordered in a variety of sizes; the size used for this project is the "12×20". This size corresponds to a height that tapers up from 12 mm at the impaction surface (i.e., the posterior surface if installed posteriorly) to 15 mm, and a mediolateral width of 10 mm maximum, and a length of

Fig. 1 A pair of Tangent allograft wedges. The white bar represents 10 mm

20 mm. These dimensions were quite variable in the samples received for this project. The Tangent has serrations on the superior and inferior surfaces. A threaded hole normal to the impaction surface allows for insertion with a threaded trocar.

Specimen preparation

Four skeletally immature juvenile bovine (calf) and four skeletally mature adult bovine (cow) cadaveric lumbar spines were obtained from slaughterhouses. The presence of cartilaginous endplates in the calf spines confirmed their immaturity, and the lack of such endplates in the cow spines confirmed their maturity. The precise ages of the animal donors are unknown, but their approximate ages are under 3 months for the calves and between 1 and 2 years for the cows. These are typical ages at which unwanted calves and beef cows arrive at slaughter. The degree of skeletal maturity for these age groups agree with the findings of a recent study on a classification system in use for feed cattle [14].

Twenty-four (12 calf and 12 cow) functional spinal units (FSUs; L1-L2, L3-L4, and L5-L6) were stripped of musculature and their neural arches removed. This was done for ease of testing and because this project only concerned vertebral body behavior. Both ends of each FSU were seated at a small depth within a polyester resin potting compound (Lite Weight 3; Fibre Glass-Evercoat; Cincinnati, Ohio). The potting allowed the FSU to be loaded in a distributed manner over the entire endplate and prevented endplate motion in the transverse plane. The calf FSUs were randomly assigned to either of two groups, such that each group contained the same number of each FSU level (e.g., each group contained two L5-L6 FSU levels). One group was the ALIF (anterior lumbar interbody fusion) group and the other the PLIF (posterior lumbar interbody fusion) group. The cow FSUs were similarly distributed randomly. One calf FSU was destroyed in preparation, leaving six calf ALIF, five calf PLIF, six cow ALIF, and six cow PLIF FSUs for testing. After nondestructive compression testing of each FSU in the intact state, bilateral Tangent pairs were implanted according to the randomization scheme, and using the surgical protocol and instrumentation set provided. Briefly, this protocol involves discectomy, distraction, decortication of the host endplates, and implantation. All procurement and use of animal tissue was conducted with the approval of the Institutional Animal Care and Use Committee.

Mechanical testing

The potted intact FSUs were placed within specially designed loading platens mounted within the operating theater of a materials

testing machine (MTS 858 Mini Bionix; MTS; Eden Prairie, Minn.). The platens allowed repeatable placement of each FSU within a horizontal plane so that the applied axial compressive loading for each FSU was at the same point in the intact and instrumented state. A rod with a spherical end, attached to the actuator of the testing machine, applied the compressive load via contact with a conformal spherical indentation in the top platen. Each intact FSU was preconditioned with ten cycles of haversinusoidal compressive loading with a period of 1 s and range of 100 N to 250 N. A haversinusoid is a sine function translated along the ordinate axis so that the minimum and maximum values are unequal in absolute magnitude and both greater than zero. This preconditioning was demonstrated in preliminary development tests to achieve a steady state condition with respect to hysteresis in the load-actuator stroke data. The intact FSUs subsequently were nondestructively

tested in compressive load control at a constant rate of 10 N/s to a maximum of 700 N. After instrumentation with Tangents, the FSUs were axially compressed to failure in displacement control at a constant rate of 0.1 mm/s. All mechanical tests were performed under slow, quasistatic loading conditions. Applied load and actuator displacement data were acquired in all tests.

Trabecular bone apparent density

After failure testing, three calf and three cow FSUs were chosen for determination of apparent density. A 15-mm-thick slice was cut from the cranial or caudal end of each FSU using a diamond-blade band saw (Gryphon; Sylmar, Calif.), then a 1-mm-thick slice was cut from the thicker slice using a precision band saw (EXAKT Apparatebau GmbH & Co. KG; Norderstedt, Germany). The slices were then cleaned of organic material through a series of two baths. The first was an immersion in a heated 50% enzymatic detergent solution (Enzol; Advanced Sterilization Products; Irvine, Calif.) for 2 h. This was followed by a polishing bath in 6% sodium hypochlorite (Publix; Lakeland, Fla.) for 15 min. The cleaned slices (Fig. 2) were then dimensionally measured using a custom area determination algorithm (MATLAB; The MathWorks; Natick, Mass.) and the thickness with a digital caliper. The area algorithm detected the perimeter of each slice using image analysis techniques to compute the enclosed area. Each hydrated slice was weighed without free water in the trabecular pores (the hydrated weight method) and apparent density ρ calculated from

$$\rho = \frac{\frac{W}{g}}{A_s t} \quad (1)$$

where W is the hydrated slice weight, g is acceleration due to gravity, A_s the slice area, and t the slice thickness. This relation gives a volumetric measure of apparent density (typical units of g/cm^3).

A literature search was conducted using several relevant keywords (e.g., lumbar, trabecular, bone, density) and their variants to

Fig. 2 Typical transverse sections through calf L5-L6 (left) and cow L1-L2 (right) vertebrae. The calf section is missing the anterolateral portions (lower left and lower right of calf image), which become separated at the uncalcified posterolateral growth plates. The white bar represents 10 mm

Fig. 3 Typical applied compressive load-actuator displacement actual data. Stiffness k is the regression line slope between 300 N and 600 N. The ultimate failure load is F_U

uncover reports on the apparent density of trabecular bone within the human adult and pediatric lumbar spine. This original search resulted in a host of papers, and their references led to even more, as is typical. The predominant method used to determine density was dual energy X-ray absorptiometry (DEXA) on living subjects. This method results in an areal measure of density (typical units of g/cm^2) rather than a volumetric measure. Since the apparent densities of the bovine vertebrae used in this project were measured directly using the hydrated weight method, a conversion from areal to volumetric density measures would have had to be performed to use these types of data. Such a conversion is not new and is discussed in papers on comparative densitometry [5, 12, 13], and usually involves a factor based on a dimensional measurement of the vertebrae in the living subject or by direct correlation on laboratory samples. The remaining papers were selected by only considering those that used the hydrated weight method or quantitative computed tomography for apparent density determination.

Data reduction and analysis

A stiffness coefficient k was defined as the slope of the regression line through the applied compressive load and actuator displacement data between 300 N and 600 N (Fig. 3) from tests of the intact FSUs and after Tangent instrumentation. These coefficients are termed the intact stiffness k_I and the Tangent stiffness k_T . The ultimate failure load F_U with the Tangent pair installed was defined as the peak load reached prior to an abrupt decrease in measured load. Otherwise, consolidation, compaction, and complete disintegration of the trabecular bone would occur if the test were allowed to proceed indefinitely, ultimately resulting in compression of the Tangents themselves. An ultimate failure strength σ_U was defined as

$$\sigma_U = \frac{F_U}{A_T} \quad (2)$$

where A_T is the combined bearing areas of the installed Tangent pairs. The bearing area of each Tangent was determined using the custom area algorithm of arrays of individual images of each Tangent. Thus, the bearing area of a Tangent pair equals the sum of two bearing areas of the individual Tangents installed in each FSU.

Statistical analyses were performed to determine the effects of the independent variables on the derived dependent variables (Table 1) and to determine distributional statistics. All statistical analyses were performed with commercially available software (StatView; Abacus Concepts, Inc.; Berkeley, Calif.). The specifics and temporal order of the analyses are detailed in the Results section. In some cases, data were pooled across independent variables after initial statistical analyses, so it is natural to present descriptions of the analyses along with the results.

Table 1 Summary of independent and dependent variables (ALIF anterior lumbar interbody fusion, PLIF posterior lumbar interbody fusion)

Variables	Factor	Level	Class
Independent variables	Maturity	Calf Cow	Nominal
	Origin of insertion	ALIF PLIF	Nominal
Dependent variables	Stiffness	Intact Tangent	Continuous
	Strength	NA ^a	Continuous
	Density	NA ^a	Continuous

Results

The results of the mechanical testing were as expected, in that the cow spines were stiffer and stronger than the calf spines (Table 2). A full interaction analysis of variance (ANOVA) was performed first to test the significance of origin of insertion, maturity, and their interaction on the intact stiffness. Origin of insertion was found not to be a significant effect ($P=0.759$) within the calf or cow group. Maturity was a highly significant effect ($P<0.0001$) on the intact stiffness. The strength of the cow PLIF group exceeded the load capability of the testing machine (4,890 N), and in all probability far exceeded the capability, as the real time load-displacement data indicated no failure was imminent. Therefore, the failure load was conservatively set equal to the machine limit load, and failure strength derived from that value. It was decided that the cow ALIF group would behave similar to the cow PLIF group, hence, the cow ALIF tests were not performed. Unpaired t -tests revealed no significant effects of origin of insertion on the calf Tangent stiffness ($P=0.426$) or strength ($P=0.311$), further supporting the decision to perform only the cow PLIF tests. The first purpose of this project was thus achieved, namely, the comparison of the compressive load bearing properties of the Tangent in the ALIF and PLIF configurations.

In light of the insignificance of origin of insertion, these data were pooled so that direct comparisons could be made between calves and cows (Table 3). Unpaired t -tests revealed highly significant differences (all $P<0.0001$) between calf and cow with respect to intact stiffness, Tangent

Table 2 Results of mechanical test data reduction split by maturity and origin of insertion

Maturity	Insertion	Stiffness ^a (N/mm)		Strength ^a , σ_U (MPa)
		Intact, k_I	Tangent, k_T	
Calf	ALIF	409±72	593±84	7.14±1.35
	PLIF	380±39	500±67	5.28±0.98
Cow	ALIF	1157±75	NT ^b	NT ^b
	PLIF	1171±76	1592±153	14.4±1.04

^a Means ± standard errors in means reported

^b NT denotes Not Tested.

Table 3 Results of mechanical test and density data reduction split by maturity and pooled across origin of insertion

Maturity	Stiffness ^a (N/mm)		Strength ^a , σ_U (MPa)	Density ^a , ρ (g/cm^3)
	Intact, k_I	Tangent, k_T		
Calf	395±41	550±54	6.29±0.87	0.424±0.027
Cow	1164±51	1592±153	14.4±1.04	0.869±0.024

^a Means ± standard errors in means reported

stiffness, and strength, with the cow being significantly stiffer and stronger. The apparent density of the cow trabecular bone was significantly greater ($P=0.0002$) than that of the calf bone. Finally, intact stiffness, Tangent stiffness, and strength all regressed significantly as power laws with density.

At least one apparent density value was found in the literature with the proper volumetric units for human pediatric and adult vertebral trabecular bone. Human adult densities were by far the most numerous. Quantitative computed tomography was used in one study [9] to determine an apparent density of 0.167 g/cm^3 for human pediatric L1-L2 vertebrae; the study population consisted of 45 pediatric patients less than 9 years of age. The hydrated weight method was used in another study [8] to determine an apparent density of 0.22 g/cm^3 for human adult L3-L4 vertebrae; the patient population consisted of 72 adults.

Discussion

The apparent densities determined in this project for bovine vertebral trabecular bone are similar to published data. The apparent density for the calf L3 vertebra has previously been reported [20] as 0.453 g/cm^3 using the hydrated weight method, compared to the mean value of 0.424 g/cm^3 found in this project. The apparent density of cow femoral condyle trabecular bone has previously been reported [4] as 0.60 g/cm^3 using the dry weight method, compared to the mean value of 0.869 g/cm^3 found in this

project. The discrepancy here is probably due to the use of the dry weight method in the previous study, and the fact that the trabecular bone came from a different region of the cow skeleton.

As previously stated, the primary purpose of this project was to establish bovine spines as models for human spines with respect to the use of Tangents in humans. To accomplish this, a Tangent usage curve was developed for the human pediatric and adult spine, as follows. The usage curves represent the maximum allowable body weight (BW) of the patient as a function of activity level so as to prevent initial failure of the host vertebral bodies. The first step in the development of these curves was to determine the power law regression between strength and density for the bovine spine data, which was found to be

$$\sigma_U = 16.891\rho^{1.1513} \quad (3)$$

Next, the human apparent densities found in the literature were substituted into the power law regression to estimate the strengths of the Tangent constructs in human spines (Fig. 4). The strengths predicted for human spines from the bovine regression compare favorably to those derived from a human trabecular bone strength-density regression of a recent study (Table 4) [11]. Then, it was assumed that the largest Tangent pairs able to be installed into a pediatric spine would be size 20, and that the size 28 Tangents would apply to the adult spine. The nominal areas for a pair of these sizes are 300 mm^2 and 400 mm^2 , respectively. Lastly, an activity factor, AF , was defined as a factor that multiplies body weight to represent the severity of various activities of life. For example, an AF of 75% cor-

Fig. 4 Power law regression on apparent density ρ and compressive strength σ_U . The calf and cow data were determined in this project. The human pediatric and adult density data are from the published literature and were substituted into the power law

Table 4 Strength estimations for human vertebral bone from bovine strength-density regression compare favorably with those from a human strength-density regression

Maturity	Apparent density ρ (g/cm ³)	Human vertebral bone strength, σ_U (MPa)		Difference
		Estimation from bovine regression (equation 3)	Estimation from human regression [11]	
Pediatric	0.167	2.15	2.15	-0.2%
Adult	0.22	2.96	3.27	+11%

Table 5 Activity factors for various activities. Subtle increases in activity cause dramatic increases in compressive loads on vertebrae

Activity	Activity factor, AF
Relaxed standing	0.75
Standing and holding 78.5 N weight close to body	1.13
Standing with trunk flexed 30° and arms extended	2.96
Rising from chair without use of hands	3.77
Standing and holding 78.5 N weight with arms extended	4.44
Standing with trunk flexed 30° and holding 78.5 N weight with arms extended	6.69
Lifting 147 N weight from floor with knees bent and arms straight down	7.06

responds to the percentage of body weight acting on the sacrum, which represents the absolute minimum compressive load on the sacrum in the absence of any muscular

activity. Simply keeping one's balance will easily drive the AF above 100%. Other values for the AF are provided, adapted from a previously published analysis of compressive loads on the lumbar spine (Table 5) [2]. The maximum allowable body weight BW as a function of AF is given by:

$$BW = \frac{\sigma_U A_T}{AF} \quad (4)$$

A graphical representation of this relation is provided (Fig. 5). From this relation, for example, an adult weighing 712 N (160 lb) should be limited to activities for which AF is less than 1.66, so as to avoid trabecular bone failure. Similarly, a pediatric recipient weighing 267 N (60 lb) should be limited to activities for which AF is less than 2.42. Neither of the activities associated with these AF s would be considered rigorous. This AF analysis would seem to indicate no bending or lifting of any weight be allowed for human patients with Tangent pairs alone, in the early postoperative period before incorporation of the grafts has occurred. Of course, this simple AF analysis does not consider the variation in trabecular bone density found in humans (although it could, simply by including such variation in the strength-density relation), nor does it consider the load sharing provided by posterior elements of the spine (primarily in ALIF procedures) or by rigid segmental instrumentation. Such instrumentation is, in fact, recommended to augment the Tangent in a decompressive PLIF procedure in the adult spine, although not specifically detailed in the brochure describing the surgical technique [16].

Fig. 5 Maximum body weight (BW) as a function of activity factor. Each curve represents the maximum BW so as to preclude trabecular bone failure of the Tangent construct

Conclusions

The Tangent allograft significantly increases the compressive stiffness of the spine using bovine spine models. The strength of the Tangent construct in the cow spine was significantly greater than that of the calf spine. A method was established to relate the results from these bovine models to estimate the strength of a Tangent construct in the human pediatric and adult spine. The method was demonstrated to indicate the appropriateness of bovine spines as models for evaluating the acute postoperative behavior of interbody allografts in human spines provided a similar methodology is used to account for the greater density and strength of the bovine vertebral bone. This method indicated that additional rigid segmental instrumentation may be necessary to preclude vertebral trabecular bone failure for all but the most limited of activities. Such instrumentation has been advocated by some for the

additional reduction in segment motion found in biomechanical studies [15], but not by all [18]. Difficulties were encountered implanting Tangents into the smaller calf segments due to the relatively long length and radius of curvature of the allografts. Further testing using Tangents with a smaller radius of curvature may be necessary to examine fully whether these design changes will significantly alter the mechanical properties of the construct, together with adjunctive instrumentation. In summary, the Tangent allograft may be suitable for use in the pediatric population, provided that an appropriate size can be manufactured, postoperative activity levels can be contained, and with the continued recommendation for adjunct instrumentation.

Acknowledgements The assistance of John Bianchi, Lukas Eiser-mann, Nils Götzen, Michelle Ortiz, and Eduardo Reyes and the financial support through a research grant by Medtronic Sofamor Danek is greatly appreciated.

References

1. Blanco JS, Sears CJ (1997) Allograft bone use during instrumentation and fusion in the treatment of adolescent idiopathic scoliosis. *Spine* 22:1338–1342
2. Bouxsein ML, Myers ER, Hayes WC (1996) Biomechanics of age-related fractures. In: Marcus R, Feldman D, Kelsey J (eds) *Osteoporosis*. Academic Press, San Diego, pp 373–393
3. Buttermann GR, Glazer PA, Bradford DS (1996) The use of bone allografts in the spine. *Clin Orthop* 324:75–85
4. Carter DR, Hayes WC (1976) Bone compressive strength: the influence of density and strain rate. *Science* 194: 1174–1176
5. Cendre E, Mitton D, Roux JP, Arlot ME, Duboeuf F, Burt-Pichat B, Rumelhart C, Peix G, Meunier PJ (1999) High-resolution computed tomography for architectural characterization of human lumbar cancellous bone: relationships with histomorphometry and biomechanics. *Osteoporos Int* 10:353–360
6. Ehrler DM, Vaccaro AR (2000) The use of allograft bone in lumbar spine surgery. *Clin Orthop* 371:38–45
7. Frost HM (1969) Tetracycline-based histological analysis of bone remodeling. *Calcif Tissue Res* 3:211–237
8. Galante J, Rostoker W, Ray RD (1970) Physical properties of trabecular bone. *Calcif Tissue Res* 5:236–246
9. Gilsanz V, Gibbens DT, Roe TF, Carlson M, Senac MO, Boechat MI, Huang HK, Schulz EE, Libanati CR, Cann CC (1988) Vertebral bone density in children: effect of puberty. *Radiology* 166: 847–850
10. Grogan DP, Kalen V, Ross TI, Guidara KJ, Pugh LI (1999) Use of allograft bone for posterior spinal fusion in idiopathic scoliosis. *Clin Orthop* 369:273–278
11. Kopperdahl DL, Keaveny TM (1998) Yield strain behavior of trabecular bone. *J Biomech* 31:601–608
12. Kroger H, Kotaniemi A, Vainio P, Alhava E (1992) Bone densitometry of the spine and femur in children by dual-energy x-ray absorptiometry. *Bone Miner* 17:75–85 [erratum 17: 429]
13. Kroger H, Vainio P, Nieminen J (1995) Estimation of spinal bone density using conventional MRI. *Acta Orthop Scand* 66:532–534
14. Lawrence TE, Whatley JD, Montgomery TH, Perino LJ (2001). A comparison of the USDA ossification-based maturity system to a system based on dentition. *J Anim Sci* 79: 1683–1690
15. Lund T, Oxland TR, Jost B, Crompton P, Grassmann S, Etter C, Nolte LP (1998) Interbody cage stabilisation in the lumbar spine: biomechanical evaluation of cage design, posterior instrumentation and bone density. *J Bone Joint Surg Br* 80:351–359
16. Medtronic (1999) Tangent posterior discectomy & grafting instrumentation set surgical technique. Medtronic, Sofamor Danek, Memphis
17. Musculoskeletal Transplant Foundation (2002) Available at: www.mtf.net
18. Ray CD (1997) Threaded titanium cages for lumbar interbody fusions. *Spine* 22:667–679
19. Regeneration Technologies Inc (2002) Available at: www.rtiotechnology.com
20. Swartz DE, Wittenberg RH, Shea M, White AA, Hayes WC (1991) Physical and mechanical properties of calf lumbar sacral trabecular bone. *J Biomech* 24:1059–1068