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Load Sharing within a Human Thoracic Vertebral Body: An In Vitro Biomechanical Study

İnsan Torasik Omurgasında Yük Dağılımı: İn Vitro Biyomekanik Çalışma

Cumhur KILINÇER[1]
Serkan İNCEOĞLU[2]
Moon Jun SOHN[3]
Lisa A FERRARA[4]
Nadi BAKIRCI[5]
Edward C. BENZEL[6]

1 Department of Neurosurgery, Trakya University Medical Faculty, Edirne, Turkey
2,4 Spine Research Laboratory, The Cleveland Clinic Spine Institute, The Cleveland Clinic Foundation, Cleveland, OH, USA
3 Department of Neurosurgery, Inje University Ilsan Paik Hospital, Korea
5 Department of Public Health, Marmara University Medical Faculty, Istanbul, Turkey
6 The Cleveland Clinic Spine Institute, The Cleveland Clinic Foundation, Cleveland, OH, USA

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Correspondence address:
Cumhur KILINÇER
Trakya Universitesi Tıp Fakultesi
Nöroşirürji AD, 22030 Edirne, Turkey
Phone: (++90) 284 2355798,
Fax: 284 2352730,
Email: ckilincer@trakya.edu.tr,
ckilincer@medscape.com

ABSTRACT

OBJECTIVE: The vertebral body is the major load bearing part of the vertebra and consists of a central trabecular core surrounded by a thin cortical shell. The aim of this in vitro biomechanical study is to determine the debated issue of load sharing in a vertebral body.

METHODS: A series of non-destructive compressive testing on excised human thoracic vertebral bodies were performed. The testing process consisted of a stepwise removal of the vertebrae's trabecular centrum and measurement of surface strains.

RESULTS: Load sharing of cortical shell of osteopenic vertebrae (48.1±7.6) was significantly higher than that of normal vertebrae (44.3±10.6). Load sharing of middle thoracic vertebrae (49.4 ±10.0) was significantly higher than that of lower thoracic vertebrae (42.4±8.5). According to general linear model analysis, test speed and load were not found to be effectual on load sharing with the exception that osteopenic vertebrae showed lower cortical load sharing under higher loads.

CONCLUSIONS: The cortical shell takes nearly 45% of physiological loads acting upon an isolated thoracic vertebra. Load sharing between cortical shell and trabecular centrum is significantly affected by spinal level and bone mineral density. The load borne by trabecular bone increases towards the lower spinal levels, and decreases by osteoporosis.

KEY WORDS: Biomechanics, Bone, Osteoporosis, Spine, Test, Thoracic vertebrae

ÖZ

AMAÇ: Omur gövdesi omurganın en önemli yük taşıyıcı parçasıdır ve merkezi trabeküler bir çekirdekten ve onu çevreleyen ince bir kortikal kabuktan oluşur. Bu in vitro biyomekanik çalışmanın amacı bir omur gövdesindeki tartışmalı yük dağılımı konusunun incelenmesidir.

YÖNTEMLER: Eksize edilmiş insan torasik omur gövdelerine bir dizi destrüktif olmayan kompresyon testi uygulandı. Test işlemi omurların trabeküler merkezlerinin adım adım çıkartılmasından ve yüzey gerilimlerinin ölçülmesinden oluşmuştur.

BULGULAR: Osteopenik omurların kortikal kabuklarının yük dağılımı (48.1± 7.6) normal omurlarınkine göre (44.3±10.6) belirgin biçimde daha yüksekti. Orta torasik omurların yük dağılımı (49.4± 10.0) alt torasik omurlarınkine göre (42.4± 8.5) belirgin biçimde daha yüksekti. Genel lineer model analizine göre osteopenik omurların yüksek yüklenmeler altında düşük kortikal yük dağılımı göstermeleri istisna olmak kaydıyla, test hızı ve yükü, yük dağılımı üzerinde etkili bulunmamıştır.

SONUÇ: Kortikal kabuk izole edilmiş bir torasik omur üzerine uygulanan fizyolojik yüklerin yaklaşık %45 ini alır. Kortikal kabuk ve trabeküler merkez arasındaki yük dağılımı spinal seviye ve kemik mineral densitesi tarafından belirgin biçimde etkilenmektedir.Trabeküler kemik tarafından taşınan yük alt spinal seviyelere doğru artmaktadır ve osteoporoz ile azalmaktadır.

ANAHTAR SÖZCÜKLER: Biyomekanik, Kemik, Osteoporoz, Spinal, Test, Torasik omurga

INTRODUCTION

Carrying the body weight is one of the fundamental functions of the spine. The load experienced by the spine is transferred from one vertebra to the adjacent one via vertebral body/intervertebral disc complex and the facet joints at the articular column. Although it is probably level- and posture-dependant, it is generally accepted that the vertebral body is the major load bearing part of the vertebra (8, 15, 20, 21). The vertebral body consists of a central trabecular core surrounded by a thin cortical shell. The load borne by a vertebral body is transmitted by way of these two paths. Thus, the overall strength of the vertebral body depends on the structural contribution and load sharing of both components (4).

What are the relative contributions of the trabecular centrum and cortical shell to load bearing capacity of a vertebra? Such knowledge has two main implications. Firstly, age-related increases in vertebral fracture incidence are generally believed to result from trabecular bone loss (3, 19, 23) and many efforts to diagnose osteoporosis have focused on monitoring bone mineral density in the centrum (7, 12, 14). However, the shell also contributes to the strength of the vertebral body, and it may play an increasingly important role in resisting vertebral fractures with aging, as central trabecular bone progressively weakens (6, 16, 28). Thus, quantification of their relative roles and relevance to age-related changes of the shell and centrum is essential to evaluate fracture risk, as well as to assess the effects of medical interventions since these therapies may affect the properties of the shell and the centrum differently. The second area for usage of the knowledge about load sharing is finite element studies. Although finite element method can be a powerful tool in studying spinal biomechanics, utilization of experimental data during modeling phase and experimental validation of the results are necessary. Thus, quantification of the load sharing in a vertebral body is of essential importance in finite element studies to achieve a correct mathematical modeling.

Despite its importance, load sharing between the centrum and shell of the vertebral body is still a debatable issue (18, 27). Rockoff et al. (24) performed nondestructive compressive tests on cadaveric lumbar vertebrae without posterior elements and found that, 45-75% of the force is transmitted via the cortical shell and that contribution of the shell increased after 40 years of age due to loss of the trabecular bone caused by osteoporosis. However, according to some authors, the cortical shell of the vertebral body, in contrast to the long tubular bones, is very thin and can only make a small contribution to its strength (2, 3, 17, 32). The results of these experiments, with methods including the removal of the cortex by manual grinding, found that the cortical shell was responsible for only 6-18% of the total compressive strength.

Several finite element analysis (FEA) studies were performed to investigate load sharing in the vertebral body (22, 26, 27, 31). Faulkner et al. (6) predicted the reduction in failure load simulating removal of the cortical shell. They estimated that the shell contributed 12% to total vertebral strength in healthy individuals and 56% in osteoporotic ones. Most FEA studies have stressed that load sharing was dependent on the vertical distance from the endplates and found that the fraction of the force taken by the shell was lower (0 to 34%) at the endplate and was higher (5 to 63%) at the mid-transverse plane, depending on the stress distribution on the disc (4, 10, 27).

The aim of this in vitro biomechanical study is to determine the debated issue of load sharing between the cortical shell and trabecular centrum in a vertebral body. Special emphasis has been placed on how such data relates to the spinal level, bone mineral density (BMD), and test conditions.

MATERIAL AND METHODS

Study design

We performed a series of non-destructive compressive testing on excised/isolated human thoracic vertebral bodies. The testing process consisted of a stepwise removal of the vertebrae's certain parts and re-test to find out the weight of each of these elements in the total load bearing capacity of the vertebra (i.e., load sharing ratios). Figure 1 and its legend explain the testing sequence and clarify the method of calculation of load sharing by means of an example. The sequence of those "remove and test" steps consisted of: 1. intact vertebral body, 2. removal of an area at the center of bottom end-plate to create a "window" in order to reach the trabecular core, 3. removal of central 25% of the trabecular bone's cross sectional area, 4.

removal of the second 25% of the trabecular bone's cross sectional area which is next to the central quarter, 5. removal of third 25% of the trabecular bone which is lateral to the second quarter, 6. removal of the last quarter of the trabecular bone which is just medial to the cortex, thus leaving the vertebra consisting of a hollowed cortical bone, like the shell of a chestnut. During all these steps, cortical strain values were measured using strain gages. Because strain is proportional to the load applied (11, 25), any increase of strain give us difference of the load borne by the (part of) vertebral body, compared to other removal steps. Thus, load sharing ratio of a given step was calculated by dividing the cortical strain value of that particular step to that of the final (hollowed) step.

Specimen preparation

Seven cadaver spines including T5-T12 levels were used for this study. Radiographs of the spines were taken to exclude vertebrae with preexisting fractures and deformities. One of the spines was used for pilot tests at the initial phase of the study, remainder six spines (45 vertebrae) were used for main study. The spines were subjected to dual energy X-ray absorptiometry (DEXA) to measure bone mineral density (BMD) in grams per square centimeter using a Hologic QDR 4500A (S/N 45451) scanner (Sterilite Corporation, Townsend, MA, USA). The spines were placed above a water bath filled with water to simulate body tissue and scanned in the anteroposterior direction. The BMD results were classified according to the T-score. T-scores greater than -1 are classified as normal, the scores between -1 and -2.5 are osteopenic, and the scores smaller than -2.5 are classified as osteoporotic. The details of the donors and specimens were given at Table I. The spines were then separated to individual forty-six vertebrae in total. Then, vertebral bodies were divided from their posterior elements by cutting their pedicles at corpus-pedicle junction using a bone saw. Surrounding soft tissues and discs of the vertebrae were removed, paying attention to keep the cortex and end plates intact.

The vertebral bodies were then partially embedded, "potted", into polyester resin (Bondo/Mar; Hyde Corporation, Atlanta, GA) in a fashion that its superior and inferior end plates filled with polyester resin and were made plano-parallel. Care was taken so that resin did not cover the cortical surfaces more than 1 mm on both ends. The specimens were wrapped in saline-soaked gauzes to prevent desiccation, sealed in plastic bags and stored at -20° C until the day of testing. On the test day, the specimens were thawed and surface preparation of the cortices of the vertebral bodies for strain gauge application was done using a procedure which was developed from the methods described by Cochran (5), and Wright and Hayes (30). This procedure involved sanding of the cortices with 220 grit sandpaper, surface cleaning with ethyl ether followed by ethanol and final neutralization with neutralizer (M-Prep Neutralizer 5A; Measurements Group, Inc. Raleigh, NC). These steps were repeated three times. After air drying the surfaces, four uniaxial strain gauges (BLH Inc., Canton, MA) were applied to anterior and lateral surfaces of the vertebral cortex in parallel to the longitudinal axis of each vertebra using cyanoacrylate, in order to find

Table I. Details of the donors and specimens tested

Specimen no.	Age	Sex	Cause of death	BMD (g/cm2)	T-score
37851	27	M	Shotgun wound to chest	1.19	0.9
41203	60	M	Cardiac arrest	1.21	1.0
43276	65	M	Myocard infarction	1.14	0.5
37858	59	M	Acute hemopericardium	1.07	-0.5
37786	55	F	Cardiopulmonary arrest	0.88	-1.5
40659	62	M	Cardiopulmonary arrest	0.81	-2.5
43138	70	F	Renal failure	0.77	-2.9

out strain distribution all over the cortex. Then lead wires were soldered to the gauges and connected to the signal conditioning equipment (Figure 2).

Biomechanical testing

The vertebrae were placed in the MTS Allience RT/10 materials testing machine (MTS Systems Corporation, Eden Prairie, MN), gripped with custom-made pincers and exposed to predetermined axial compressive load which detailed below. The vertebrae were cycled six times with combinations of different loads (200-400-600 N) and crosshead displacement speeds (1-5-10-25 mm/sec). Thus, each vertebra was tested twelve times at each step. After intact testing, vertebral body was separated from the bottom fixture, and a window was created at the middle of the bottom end-plate using electric drill, without destructing underlying trabecular bone. After testing of end-plate removal (EPR) step in the same manner, the trabecular bone was removed in a stepwise fashion which described above (see legend of Figure 1). To ensure that the proper amount of trabecular bone was removed, two methods were used consecutively: first, the cross-sectional areas of specimens were calculated using digital photography and an image-analysis program (Scion Image for Windows, Scion Corporation, Frederick, Maryland, USA) to determine the radius of trabecular bone area which is about to be removed

Figure 2: A view of the experiment design. The wires protruded from the specimen's strain gages were connected to the signal conditioning equipment.

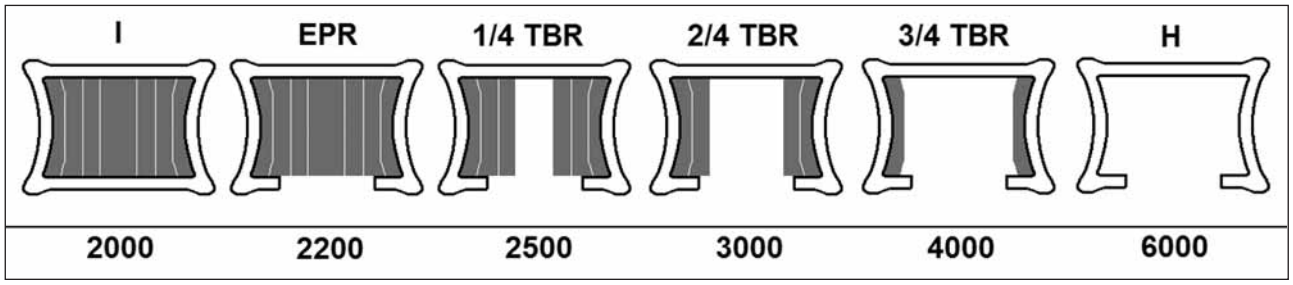

Figure 1: The testing sequence the method of calculation of load sharing (I= Intact, EPR= End-plate removal, TBR= Trabecular bone removal, H= Hollowed). Each vertebral body was tested in six steps: intact, after end-plate removal, and after stepwise removal of each quarter of the trabecular bone beginning with its center. The numbers at the bottom are an example set for cortical strains. These are not the actual results and just given to explain the method of calculation of load sharing. In this example, cortical strain of intact vertebra was given as 2000 and that of hollowed vertebra was 6000. Thus, considering the difference between intact and hollowed vertebrae is the lack of trabecular bone, the trabecular bone should be responsible for the difference between strains, which was 4000. Strain is proportional to the load borne. Thus, when we consider the intact vertebra, the fraction of strain (thus, load) borne by cortex in the total strain gives us the load sharing ratio of the cortex, which was 2000/6000 = 1/3, i.e., 33%. In the same way, when we consider EPR step, 2200/6000 = 0.37 (37%) gives us the load borne by the cortex plus bottom end-plate; 2500/6000 = 0.42 (42%) gives the load borne by cortex plus bottom end-plate plus central quarter of the trabecular bone; etc. Serial calculations and subtractions give the relative contributions of each part to the total load borne by the vertebra, which are 33% cortex, 4% bottom end-plate, 5% central quarter of the TB, 8% second quarter of the TB, 17% third quarter of the TB, and 33% by the fourth (outermost) quarter of TB, in this example.

(Figure 3). Secondly, care was taken to remove equal amount of bone during each step of removal. In order to do this, drilled bone was collected and its weight was measured using a digital weight meter with a resolution of 0.01 gr. Thus, precise removal of each quarter of the trabecular bone (TB) was ensured. After each step of TB removal process, the vertebra was tested in the same manner (at different loads and different speeds). Because each vertebra was tested twelve times at each step, there were totally 72 tests for each vertebra.

Figure 3: The bottom view of the vertebra after creation of a window at the center of endplate to reach trabecular bone. Note that an area of the central 25% of the trabecular bone was determined and carved.

Collection of data

During the loading, microstrains were continuously measured and recorded at a rate of 50 Hz using a data acquisition system consisting of a strain gauge card, a scanner, a peripheral connect interface card (Model 5110 Strain Gauge cards, Scanner Model 5100A, PCI Interface card model 5101A, Vishay Micromeasurements Group, Inc., Raleigh, NC) and a personal computer (Dell Dimension 4300 PIV, Dell Computer Corporation, Round Rock, TX, USA). Compressive strain is the percent decrease in length (negative sign), and tensile strain is the percent increase in length (positive sign). Except a few cases, the gages returned compressive strains. The total strain of each test was defined as the sum of the absolute values of the tensile and compressive microstrains obtained from four gauges. The strain data of the sixth cycle were sampled from each of the four gauges using the system's software (StrainSmart, Ver 2.23, Vishay Micromeasurements Group Inc., Raleigh, NC) and exported into Microsoft Excel 2000 (Microsoft Of?ce Excel 2000; Microsoft, Redmond, WA) for processing before final import into the statistical program. Other than the strain data, the load-displacement data acquired by the testing apparatus (MTS Alliance RT/10) were recorded at a sampling rate of 50 Hz.

Analysis of data

In order to show the how trabecular bone removal affects strain and stress values, and effect of age, level, osteoporosis and testing conditions on the load sharing; descriptive statistics and inferential statistical tests were employed using a statistical software program (SPSS 11.5, SPSS Inc., Chicago, IL). General Linear Model (GLM) for repeated measures in different speeds and loads was used for the analysis of variance. Osteopenia (osteopenic or normal) and the level of the thoracic vertebra (middle or lower) were defined as the between-subjects factors in the model. The difference of cortical strains or LS ratios in the intact vs the EPR mode was analyzed by paired T test. Student's t-test was used to compare the means of cortical strain or LS ratios in osteopenic and normal vertebrae, middle and lower vertebrae groups. The tests were performed at the 95% confidence level.

RESULTS

Cortical strains at the intact mode

Table II shows cortical strains of the middle and lower thoracic levels of normal (non-osteopenic) and osteopenic intact vertebrae in every loads and test speeds. In the middle thoracic region, osteopenic vertebrae showed nearly two times higher strains compared to non-osteopenic vertebrae, in each test speed (1-5-10-25 mm/sec) and each load (200-400-600N) ($P<0.0001$ for each). The difference between osteopenic and normal vertebrae in the lower thoracic region was modest, but was still significant ($P<0.05$). Strains of the normal vertebrae showed a small inter-region (i.e., middle and lower thoracic) difference, and this difference insignificant. However, strains of middle thoracic osteopenic vertebrae were significantly higher than those of lower thoracics ($P<0.05$). Strains linearly raised under increasing loads. When all strain data were pooled (disregarding existence of osteopenia and other variables), mean strain was 1156 ± 578 at 200N, 2347 ± 964 at 400N, and 3376 ± 1408 at 600N load.

Table II. Cortical strains at the intact mode (Mean±SD)

Test speed (mm/sec)		1			5			10			25			MEAN	
Load (N)		200	400	600	200	400	600	200	400	600	200	400	600		
Middle Thoracic	Normal	929	1937	3014	872	1831	2818	878	1810	2794	976	1931	2865	1888	
		±350	±752	±1062	±329	±727	±1046	±335	±720	±1064	±396	±803	±1105	±1089	2489
	Osteopenic	2087	3956	5512	1994	3840	5458	2008	3818	5427	2360	4305	5931	3891	±1597
		±566	±1104	±1462	±628	±1099	±1439	±643	±1067	±1411	±824	±1177	±1567	±1727	
Lower Thoracic	Normal	1012	1970	2898	935	1853	2767	956	1850	2773	1087	2018	2954	1923	
		±442	±879	±1184	±440	±805	±1132	±460	±840	±1181	±548	±929	±1242	±1137	2088
	Osteopenic	1179	2526	3862	1150	2435	3805	1261	2588	3980	1380	2751	4080	2583	±1182
		±326	±520	±608	±318	±464	±520	±455	±655	±634	±370	±478	±454	±1191	
MEAN		1155	2306	3423	1089	2196	3292	1116	2207	3306	1265	2402	3485		
		±548	±1043	±1386	±543	±1011	±1387	±564	±1032	±1414	±672	±1157	±1531		
			2295			2192			2210			2384			2270
			±1392			±1367			±1380			±1473			±1398

Thus, strain positively correlated with load, as expected. Test speed did not affect strains, according to the general linear model analysis.

Load sharing after trabecular bone removal

As the trabecular bone was removed step by step (intact, end plate, 1/4 TB, 2/4 TB, 3/4 TB, and whole TB), the strains of the cortex increased gradually, indicating the increased load borne by the cortex. According to the method described in the previous section, strain values were converted to percent of load sharing of the cortex. The differences between load sharing ratios of other removal steps were significant at each test speed and load ($P<0.0001$ for each step). General linear model analysis showed that this finding was independent from osteopenia and level. Mean ± standard deviations of the cortical load sharing after each step of the removal process at 1mm/sec test speed under 400N load is presented in Table III. The results of the tests performed at other speeds and loads were similar. While the amount of increase of load sharing was relatively small (8-10%) after removal of first two quarters (middle of the vertebrae), it was substantial (nearly 50%) after outer two quarters of the trabecular bone vanished (Figure 4).

Table III. Load sharing of cortical shell after stepwise removal of trabecular bone

	Mean (%)	Std. Dev.
Intact	42,8	9,4
End plate removal	45,7	9,8
1/4 TB* removal	48,3	9,5
2/4 TB removal	52,8	9,0
3/4 TB removal	64,9	9,5
Total TB removal	100,0	0,0

Test was performed at 1 mm/sec test speed under 400N (n=45), *TB: Trabecular bone

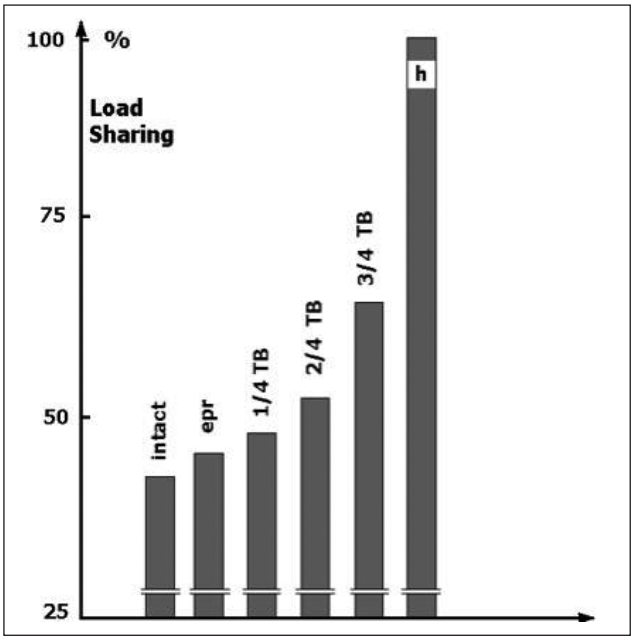

Figure 4: Cortical strains after stepwise removal of trabecular centrum. While the amount of increase of load sharing was relatively small after removal of first two quarters (middle of the vertebrae: 1/4 TB and 2/4 TB), it was substantial after outer two quarters of the trabecular bone vanished (3/4 TB and h step). (epr: end-plate removal, TB: trabecular bone, h: hollowed, total trabecular bone removal).

Factors affecting load sharing of cortical bone

At this point of data analysis, load sharing ratio of EPR step was taken as cortical load sharing of a given vertebra. (Table IV) shows load sharing at each test speed for osteopenic and normal vertebrae. Certain factors such as the level, existence of osteopenia, test speed and load were analyzed using general linear model to show their effects on cortical load sharing. Cortical load sharing of osteopenic vertebrae was higher than those of normal vertebrae, irrespective of test speed and load. When all the data were pooled, cortical load sharing of osteopenic vertebrae was 48.1±7.6 and was 44.3±10.6 for the normal vertebrae, a statistically significant difference (p=0.03). Likewise, cortical load sharing of middle thoracic vertebrae (49.4 ±10.0) was significantly higher than those of lower thoracic vertebrae (42.4±8.5, p=0.05). According to general linear model analysis, test speed and load were not found to be effectual on load sharing with the exception that osteopenic vertebrae showed lower cortical load sharing ratios under higher loads (P=0.04). (Figure 5) shows the relationship between load sharing and test load.

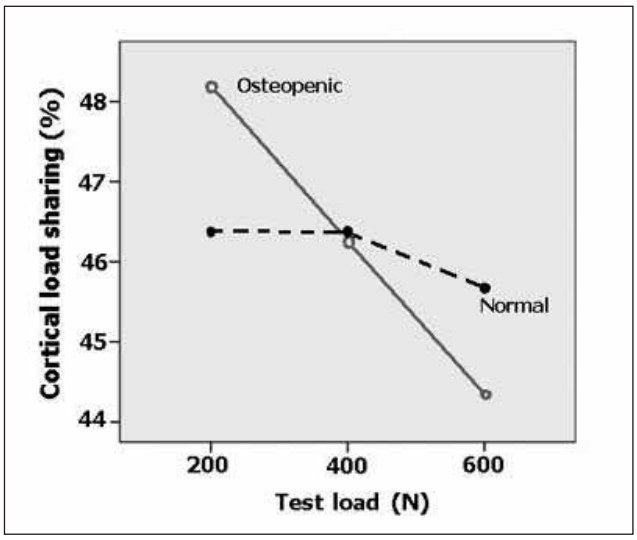

Figure 5: The relationship between load sharing and test load. The osteopenic vertebrae showed significantly lower cortical load sharing ratios as the load increased. The test was performed at 1 mm/sec test speed.

DISCUSSION

In situ determination of relative structural roles of the shell and centrum is difficult because there is no way to test each component's weight in a morphologically intact vertebra. Instead, studies tried to remove one of the components, to test the remainder component, and compare the result with the intact one to find out the weight of the removed part. Previous experimental studies had used peak load or load/displacement curve to calculate load sharing ratios of cortical shell and trabecular core. The utilization of the strain gauge technology was the main difference of our experiments from previous studies. The logic behind utilization of cortical strains was the fact that applied load and strain were correlated linearly. In fact, during our experiments, load-strain curves showed a linear relationship. Strain linearly increased 1000-1200 for

Table IV. Load sharing ratio of cortical shell at the EPR mode (% mean ± sd)

Test speed (mm/sec)		1			5			10			25			MEAN	
Load (N)		200	400	600	200	400	600	200	400	600	200	400	600		
Middle Thoracic	Normal	49.4 ±13.2	45.8 ±11.7	51.8 ±9.7	50.1 ±13.3	45.4 ±12.7	51.2 ±10.5	51.3 ±14.1	45.0 ±12.4	50.7 ±10.2	51.0 ±15.5	50.5 ±12.7	50.5 ±9.7	48.9 ±11.8	49.4 ±10.0
	Osteopenic	51.2 ±7.5	50.6 ±5.7	45.0 ±7.9	49.1 ±4.9	51.5 ±6.0	46.5 ±7.0	51.3 ±5.1	52.5 ±5.6	46.2 ±6.7	48.7 ±3.4	50.2 ±5.0	48.4 ±7.9	50.6±5.8	
Lower Thoracic	Normal	43.3 ±11.0	41.2 ±6.7	39.9 ±7.0	40.9 ±8.9	40.2 ±6.8	40.0 ±7.4	40.9 ±9.0	40.0 ±7.0	39.8 ±7.6	41.7 ±9.9	40.5 ±9.2	40.2 ±8.3	40.7±7.8	42.4 ±8.5
	Osteopenic	45.2 ±9.2	46.3 ±10.6	43.7 ±4.4	44.7 ±8.6	46.8 ±11.6	44.2 ±4.2	46.3 ±8.5	47.0 ±11.7	45.0 ±4.5	43.7 ±5.4	46.5 ±3.9	46.6 ±3.3	45.9±8.7	
MEAN		46.6 ±11.0	45.7 ±9.2	44.9 ±9.0	45.5 ±10.4	45.7 ±10.1	45.0 ±9.1	46.3 ±11.0	45.8 ±10.2	44.9 ±9.0	45.9 ±11.3	45.8 ±10.2	45.5 ±9.1		45.7 ±9.8
		45.8±9.5			45.5±9.9			45.7±10.1			45.7±10.1				

each 200N of load, confirming the idea behind our experiment design. Thus, measuring the cortical strains enabled us to determine the load transferred from the cortex. This fact, in turn, brought on the second idea utilized in the current study: by dividing a hollowed vertebra's cortical strain to that of its intact situation gave us the proportion of the cortical strain of its intact form's strain, thus, the load shared by the cortex.

Most of the previous experimental studies' design on load sharing had included removal of the cortex by grinding and studying on the trabecular bone. However, since we decided to measure cortical strain, the only way to study on load sharing was to remove trabecular bone. At the initial phases of the study, a series of demo tests were performed in order to find out the most appropriate way of trabecular bone removal without significantly affecting the integrity of the cortical shell. Before we decided to the current method, a couple of different methods of TB removal were tried. For example, vertebral body was cut transversely, dividing it upper and bottom halves. After TB removal, the two parts were matched taken to original position, glued with cyanoacrylate in order to prevent slippage, and tested. The results were not consistent and this method was abandoned. Secondly, the Rockoff method (24) was tried: the integrity of trabecular bone was disrupted horizontally using the drill introduced into the vertebra, through the basivertebral vein canal. However, we found that disrupting the TB without endangering the cortical shell was difficult, and in some areas, there was some trabecular bone neighboring the cortex left. This resulted in a small amount of strain increase suggesting that the remainder of the trabecular bone still contributed to load transmitting. In fact, when more TB removal was performed by entering through pedicles, surface strains increased considerably. Thus, the Rockoff method was thought to be inappropriate, at least when load sharing was determined using strain gages. In their FEA study, Mizrahi et al. found that (18), inferior cortical shell did not bend, although experienced considerable tensile stresses under axial loading. This finding made us think that a limited window created in the middle of the inferior endplate (which we called endplate removal, EPR) might not affect load bearing characteristics of the vertebra considerably. In fact, our demo tests showed that EPR changed the surface strains little. Our results confirm that this procedure enabled us to reach TB without changing specimens' load bearing properties significantly.

An important factor which adds to complexity of the load sharing is the variability of the ratio depending on location in a vertebra. None of the previous experimental studies had mentioned such a location-dependant LS ratio. However, although their numbers varied in a wide range, most of FEA studies agree with the cortical shell force was the lowest at either endplate (0% to 34%) and reaches to its maximum at the midtransverse plane (5% to 63%), thus reporting a single LS value was inappropriate (4, 10, 27). Therefore, in our preliminary tests, after decision of the method of TB removal, we tried to investigate strain distribution all over the cortex in order to validate location-dependant LS ratio hypothesis of FE studies, and to decide where to apply strain gauges. Two T12 vertebrae, which their surfaces were large enough for a precise strain mapping, equipped with strain gages all over their anterior and lateral cortex (using 17 gauges) and subjected to the same test procedure used in this study. Because the data obtained from these specimens were limited, their results were not detailed here. However, the results gave us a clear opinion about distribution of load sharing. The results of these two specimens showed that increment of surface strains by trabecular bone removal, which gave the load sharing of cortical bone, was the most prominent at the middle of the vertebrae and was the lowest (nearly 0%) near the endplates. These data were in line with the results of FEA studies. Because the gauges which were applied close to bottom endplate gave inconsistent results (at some gauges, strain decreased after TB removal), we thought that experiment design supply us more consistent results at the mid-sagittal point (the waist) of the vertebrae.

We found that at the midsagittal point, the cortical bone took nearly 45% of the total axial load acting upon an isolated thoracic vertebra. However, our data also showed that load sharing in a vertebral body was affected by the certain factors, and reporting a single value was misleading. For example, we found that BMD was effectual on load sharing. The literature regarding the effect of osteoporosis on load sharing is conflicting. Most of authors suggested the contribution of the cortical shell increased as BMD decreased. With aging,

trabecular bone is lost at a higher rate than the cortical ring (13, 23, 29) and the cortical ring contributes with an almost constant absolute but increasing relative value to the total vertebral body strength (19). Faulkner et al. (6) estimated that the shell contributed 12% to total vertebral strength in healthy individuals and 56% in osteoporotic ones. Contrary to other studies, Homminga reported that contributions of the cortical and trabecular bones to the total load transfer do not differ between healthy and osteoporotic vertebrae. Our results show that, compared to normal vertebrae, load sharing of the cortical shell was higher in osteoporotic vertebrae. However, considering previous reports, the difference between normal and osteoporotic vertebrae was quite small, and was especially prominent for lower thoracic region. Thus, it seems that the effect of osteoporosis on load sharing is level-dependant. As the volume of the trabecular bone increased (as it was observed in the lower thoracic levels), the effect of osteoporosis tended to be more obvious. From the finding of the current study, we suggest that as the BMD decrease, the contribution of the shell to the total compressive strength increases. However, to observe this effect, enough trabecular bone volume may be required and it probably could be observed in the lower thoracic and probably lumbar levels.

We found that spinal level influenced load sharing. This effect was significant for both normal and osteopenic vertebrae. When all data were pooled, the load sharing ratio of cortical bone of middle thoracic vertebrae was 7% higher than those of lower thoracics. In conclusion, the load sharing between cortical shell and trabecular centrum was significantly affected by spinal level and BMD. We therefore suggest that because the percent volume of the TB increases in large vertebral bodies (like lower thoracics and lumbar levels), the load borne by TB increases towards the lower spinal levels. Also, the effect of osteoporosis on load sharing (which decreases the load borne by TB) becomes more prominent in lower regions.

There is substantial evidence from in vitro testing of vertebral bodies in uniform compression that fracture involves either end plate of cortical shell (9, 17). Increased cortical strains can indicate whether and where was the cortical bone was under fracture risk. Thus, we analyzed our data to find out which factors are effectual on cortical strain. We found that cortical strains were significantly affected by osteopenia and spinal level. Under the physiological load we applied, osteopenic vertebrae (which should have a higher risk of fracture) showed considerably higher strain values compared to normal vertebrae. The cortical strains of the osteoporotic vertebrae were also level-dependant: smaller vertebrae (middle thoracics) had significantly higher strains than those of the lower thoracic levels. Non-osteopenic vertebrae did not show a relationship between spinal level and cortical strains. The difference of osteoporotic vertebrae of middle and lower thoracic regions may require increased incidence of osteoporotic fractures at the middle thoracic region compared with lower thoracic levels. However, under in vivo conditions, this difference can be neutralized by the fact that upper levels are subjected to less load and maybe the contribution of rib cage is more prominent at middle thoracic than that of the lower thoracic region.

In similar biomechanical experiments, 1 mm/sec loading rate is a commonly used value, which many authors preferred. By repeating all the test at different loading speeds (1, 5, 10, and 25 mm/sec), we studied the effect of test conditions on the results. We found no significant difference between results of different speeds neither on the cortical strains, nor on the load sharing. This finding suggests that faster speeds are acceptable in similar experiments. Of note that even though we studied 25 mm/sec loading rate, our results reflect the conditions of physiological loading encountered during daily life. It remains still unclear what is the load sharing between cortical and trabecular bone in high-velocity trauma conditions such as burst fractures. This trauma models requires much faster loading rates and their fracture pattern are different than that of the osteoporotic fracture. Thus, our results are applicable to physiological loads subjected in daily life and in osteoporotic fractures.

Instead of removal of the whole TB, we preferred a gradual bone removal starting from center of the TB, in order to obtain additional data. We found that removal of inner half of TB did not increase cortical strains too much, indicating the relatively low contribution to load borne by the vertebra of this part of bone. However, the contribution of outer half of the TB, especially the outermost quarter, to the load sharing was substantial. Considering that osteoporosis affects TB starting from the center of the

vertebra (19), our experiment design may give some insights about the effect of osteoporosis on load bearing ability of the vertebrae. We suggest that at the beginning of the process, osteoporosis may affect the total compressive strength very little, although the total bone mass decreased considerably. Then, the strength of the TB decreases rapidly when the effect of the osteoporosis reached to outer half of the TB, despite a low change in BMD. It is of note that the total shares of the cortex and outer half of the TB reaches to 90% of the load. Thus, when we consider load bearing parts of a vertebra at the mid-sagittal plane, vertebral body resembles a pipe, rather than a homogenously solid structure.

The findings obtained by the current study reflect load sharing ratios of cortical shell and trabecular core under physiological loads. The ratios might be different at supra-physiological loads. For example, the contribution of the trabecular bone to the total strength may be small at lower physiologic loads, may increase towards to the limit of that vertebra can bear, and may reach its maximum just before the vertebra fractured. Thus, rather than being a constant ratio, load sharing in a vertebra may be a function of the amount of load borne. Previous experimental studies (17, 24, 32) did not concern about this issue, and mainly concentrated on the load sharing at the failure point of the vertebral body, not at the lower loads. The current study investigated the load sharing ratio at different loads (200 to 600 N), in order to examine this problem. 600 N was the upper limit of our experiment design may allow, because our preliminary test showed that some hollowed vertebrae of middle thoracic region was fractured above that level. When we considered the whole series, we found that there was no relationship between load sharing and the applied load. However, depending on BMD and spinal region, this relation may occur partly. We found a relationship between load sharing and applied load in osteopenic vertebrae especially of middle thoracic region: in these vertebrae, cortical load sharing at 600 N was significantly lower than those of 200 N and 400 N. In other words, in small and osteopenic vertebrae, towards the limits of the load those vertebrae could bear, trabecular bone's share increase. This may be caused by unresponsiveness of the cortex strain to further TB removal, since maximum deformation (strain) has already occurred. Such a finding did not exist in normal vertebrae and lower thoracics, maybe because of the fact that we did not applied enough load to reach their maximum loading capacity. Thus, we need to stress that the load sharing ratio between cortical and trabecular bones found in the current study reflect the values valid for physiological loads only, not the values valid for failure point. Because many osteoporotic fractures occur at the physiological loads, our load sharing numbers should be useful in clinical usage, given that knowing the share of the TB may increase some more towards the failure load.

The current study has several limitations which deserve to be mentioned. In this in vitro study we will test isolated vertebral segments in an artificial environment, removing their connections with surrounding tissues (intervertebral discs, adjacent vertebrae, ligaments, facet articulations, and muscles). Other than over-simplifying effect of ex vivo conditions, integrated structure of the cortical and TB adds difficulty to estimate load sharing. When determining the load-bearing ratios of individual bone compartments, it is imperative to take into account the complex interaction between the spongiosa and cortical bone and their mechanical bond in the vertebral body (1). We did not examine load sharing in a morphologically intact vertebra, the same limitation which all previous experimental studies suffered. Our experiment design included creation of a window at the bottom endplate to reach TB. However, we found this method was the least harmful method in terms of disrupting normal load-bearing function, and its effect was minor.

CONCLUSIONS

The results of the current study suggest that load sharing in a vertebral body is a complex issue and numerous factors are effectual on load sharing between vertebral elements. Cortical shell takes nearly 45% of physiological loads acting upon an isolated thoracic vertebra. The load sharing between cortical shell and trabecular centrum is significantly affected by spinal level and BMD. The load borne by trabecular bone increases towards the lower spinal levels, and decreases by osteoporosis.

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