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Mineral content and strength of lumbar vertebrae

A cadaver study

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Fifty-two cadaveric spine-motion segments were tested in compression alone and in combined compression-flexion to determine whether the compressive strength of lumbar vertebrae varied with the direction of the applied load, that is, whether similar relationships existed between the compressive strength and the amount of bone mineral depending on the direction of the loading.

The bone mineral content (BMC) ranged between 1.6 and 5.8 g/cm and the ultimate strength between 810 and 10,090 N. The BMC of the motion segments was correlated with their strength irrespective of degree of flexion during testing (0–15°). For compression-flexion within physiologic limits, the first part of the motion segment to fail was, with few exceptions, the end plate and the adjacent spongy bone.

A close positive linear relationship has been found between the axial compressive strength and the amount of bone mineral in human lumbar vertebrae (Bartley et al. 1966, Bell et al. 1967, Galante et al. 1970, Hansson et al. 1980, Lindahl 1976). Comparisons between in vivo loads and results obtained during strength testing in vitro have clearly demonstrated how close to a failure relatively normal activities and exertions bring the human lumbar spine (Andersson et al. 1980, Hansson et al. 1984, Schultz et al. 1982). One explanation for this narrowness between normal and "abnormal" could be that most load calculations in vivo have been performed while lifting, that is, in most cases with the lumbar spine more or less flexed. The in vitro strength-testing experiments, on the other hand, have usually been performed while the compressive load has been applied axially. With the highest loads in vivo during flexion, it seemed reasonable to assume that the vertebral bodies would adapt and remodel accordingly. The purpose of our study was to find out if special relationships existed between the compressive strength and the bone mineral content when the compressive load was applied axially or in different degrees of flexion.

Materials and methods

Twenty-one fresh, intact spine sections, including the vertebrae from T12 to S1, were excised during routine autopsies. Fifteen spines were from male subjects, with a median age of 31 (17–75) years, and six were from female subjects, with a median age of 70 (35–78) years. All the subjects had died from acute diseases or injuries (Table 1) and had spent less than 2 weeks in bed prior to death. Included were only spines from subjects without known malignancies or disorders that could affect the normal metabolism of the bone tissue.

All the tissues surrounding the spine were excised except the spinal ligaments and the intervertebral disks. After the excision the spines were divided into 52 motion segments, which consisted of two vertebrae and their intervening disk. To exclude specimens with abnormalities, such as metastases, fractures, or pronounced degenerative changes, lateral and frontal plane radiographs were taken of each segment. When stored the specimens were always kept in sealed plastic bags to preclude drying. In between the different tests, the motion segments were kept frozen at –22 °C.

Bone mineral determination

Dual photon absorptiometry was used to determine the bone mineral content (BMC) in the lumbar-spine motion segments (Roos 1975, Roos and Sköldbom 1974). Our absorptiometric method uses two radionuclides, which both emit gamma radiation at different energies (americium-241 and cesium-137). The radiation sources are positioned so that there is a common collimated radiation beam. This beam is projected at the

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Table 1. Bone mineral content and strength of cadaver lumbar vertebrae

A	B	C	D	E	F	G	H	I	J
1	1	74	F	L 1–2	1	2.806	2930	0	I
	2			L 3–4		3.009	4232	0	I
2	3	69	F	L 2–3	2	4.110	3582	0	I
	4			L 3–4		3.288	4168	0	I
3	5	69	F	T 12–L 1	2	2.123	2280	0	I
	6			L 2–3		2.536	1628	0	I
	7			L 4–5		3.123	3256	0	I
4	34	18	M	T 12–L 1	3	2.940	6520	10°	II
	8			L 2–3		3.701	7652	0	I
	35			L 4–5		3.836	5535	10°	II
5	41	24	M	T 12–L 1	3	4.330	5828	15°	II
	42			L 2–3		4.519	6024	15°	II
	9			L 4–5		5.063	9116	0	O
6	10	34	M	T 12–L1	4	2.527	4884	0	I
	11			L 2–3		3.410	6024	0	I
	12			L 4–5		2.951	6184	0	O
7	40	35	F	T 12–L 1	3	2.645	3907	10°	I
	13			L 2–3		3.218	5210	0	I
	14			L 4–5		3.314	5291	0	I
8	33	20	M	L 1–2	5	4.270	8303	10°	I
	15			L 3–4		4.992	10090	0	I
9	16	71	M	T 12–L1	1	2.516	3289	0	O
	17			L 2–3		2.790	4314	0	I
	18			L 4–5		2.761	4558	0	I
10	19	78	F	T 12–L 1	1	1.662	1660	0	I
	49			L 2–3	1	1.746	810	15°	II
	50			L 4–5		1.954	1296	15°	II
11	20	75	M	T 12–L 1	1	2.123	4396	5°	I
	21			L 2–3		2.536	5700	5°	I
	22			L 4–5		3.123	5210	5°	O
12	23	78	F	L 1–2	6	1.835	1628	5°	I
	24			L 3–4		1.715	1620	5°	I
13	25	34	M	L 2–3	5	4.675	5861	5°	O
	26			L 4–5		3.361	8742	5°	I
14	27	41	M	T 12–L 1	7	3.540	5551	5°	I
	28			L 2–3		3.915	6675	5°	I
	29			L 4–5		4.325	8547	5°	O
15	30	17	M	L 4–5	8	2.853	5047	5°	I
16	31	26	M	L 2–3	9	3.512	6512	5°	I
	32			L 3–4		4.304	9556	5°	I
17	36	33	M	T 12–L 1	5	3.950	5047	10°	II
	37			L 2–3		4.201	7814	10°	I
	48			L 4–5		5.262	5454	15°	II
18	38	31	M	L 1–2	3	5.052	7814	10°	II
	39			L 3–4		5.145	8433	10°	II
19	43	20	M	T 12–L 1	3	3.991	6512	15°	O
	44			L 2–3		4.745	7814	15°	II
	45			L 3–4		4.990	7652	15°	II
20	46	28	M	L 1–2	8	4.993	7000	15°	II
	47			L 3–4		5.854	8547	15°	II
21	51	34	M	L 2–3	8	3.590	3078	15°	II
	52			L 4–5		3.411	3677	15°	II

A subject number
B specimen number
C age
D sex
E segment

F cause of death
  1 myocardial infarction
  2 cerebral hemorrhage
  3 intoxication
  4 C2-fracture
  5 cerebral contusion
  6 insufficientia cordis
  7 gunshot
  8 acute respiratory failure
  9 aortic rupture

G BMC g/cm
H ultimate compressive strength (N)
I amount of flexion
J type of fracture (see text).

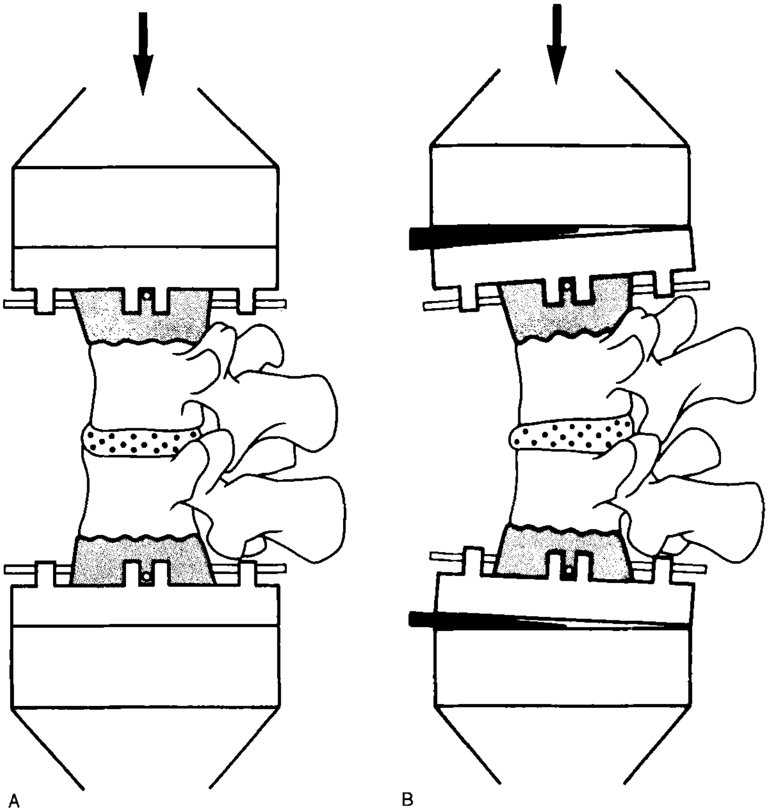

Figure 1. Compression test.
A. The motion segment between the cross heads of the testing apparatus, the bone cement, and the Steinmann pins driven through the bone cement to enhance the fixation stability.
B. The motion segment during flexion and simultaneous compression. The flexion was achieved through exchangeable wedges.

center of each vertebra using an x-ray tube and an image intensifier. The lumbar motion segment, which is submerged in water during the measurements, is moved in the transverse direction in 4-mm steps. Transmission measurements are made between the steps and during preselected time intervals. The photon energies are exponentially attenuated by the vertebra. The data are computerized to give the BMC in grams per centimeter.

Mechanical testing

The proximal and distal end plates of each segment were removed through parallel cuts. The raw bone surfaces were then embedded in a two-component plastic cement (Plastic Padding Hard®). Before the cement had set, two steel Steinmann pins were driven through the bone and also embedded in the resin at each end of the segment (Figure 1) to allow flexion in one interspace and exclude all other motions during testing. To enhance the contact between the cement and the trabecular bone, slight compression was applied during setting. The final setting of the cement took place between parallel surfaces.

All the specimens were tested in axial compression alone or in different degrees of flexion and simultaneous compression using an MFL EZU-100 dynamic testing apparatus. Load and deformation were recorded on an x-y plotter. Exchangeable wedges allowed compression-flexion in the intervertebral joint of 5, 10, or 15°, respectively. The compression took place at a constant speed of 12 mm/min. The load was stopped immediately after the first sign of failure within the segment. A failure was noted as a sudden increase in deformation on the load-deformation curve. This increase in deformation was in several cases accompanied by an audible crack and often by blood and bone marrow being squeezed out of the vertebral body.

Examination after testing

New radiographs of the segments in the frontal and lateral projections were taken after the mechanical testing. The posterior elements were then removed by a cut through the pedicles. The posterior part of the vertebral bodies and the intervening disk were examined for fractures or herniations of the disk. Because the segments were sliced to reveal the type of failure, it was not possible to grade the disk degeneration. Each segment was then cut sagitally into 3–4-mm-thick sections. After mechanical cleansing in a fine jet of water, the slices were defatted in alcohol and bleached in 10 percent hydrogen peroxide. Each slice and intervening disk were then scrutinized under 10–15x magnification for failures caused by the mechanical testing. The preparation of the segments after testing made grading of the degree of disk degeneration impossible.

Conventional statistical methods were used, and statistical significances were tested at the 5 percent level.

Results

Two different types of microscopic fractures were observed.

Type 1. A fracture located in the center of the end plate. An intact segment of the end plate had been pressed into the underlying spongy bone. Trabecular fractures surrounded the fracture segment.

Type 2. A wedge fracture with compression of the anterior part of the vertebra from just below the cement fixation. This type of fracture was only seen after compression in flexion.

Microscopic fractures were found in 45 of the 52 specimens. Twenty-nine segments had a type 1 fracture, 16 had a type 2 fracture, while no fractures were detectable in seven of the specimens, although signs of failure were recorded in the load-deformation curve (Table 1).

There were no differences in the distribution of age, sex, bone mineral content, or spinal level of the tested segments when specimens with and without fractures were compared.

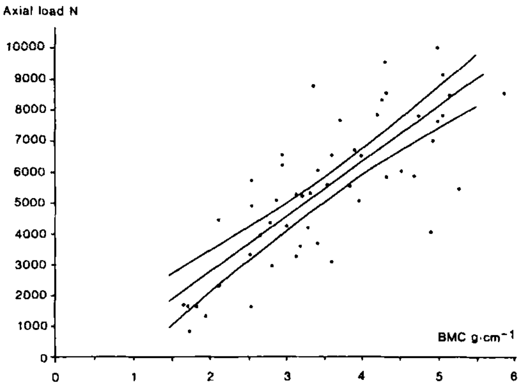

Figure 3. The relationship between ultimate compressive strength and bone mineral content in 52 motion segments. The regression line is defined by the following equation: Axial load = 1685 x BMC −322. The slope of the line is positive and statistically significant (*P*< 0.0001).

The load-deformation curves indicated that the tested segments failed in mainly two different ways. After a typical elastic deformation and a short plastic deformation, the load-displacement curves in some specimens suddenly showed a sharp deflection. This sudden drop was often accompanied by an audible crack. In other specimens, there was no such sharp deflection, but instead a smooth gradual deflection of the load-displacement curve, often accompanied by a more continuous crushing sound. It was not possible to relate the different failure patterns to a certain type of fracture or nonfracture.

When all the tested segments, irrespective of the direction of loading, were tested as one single group, a correlation was found between the BMC of the specimens and their ultimate compressive strength (Figure 3).

When the specimens were grouped according to the degree of flexion during the compression test and the same relationship was tested in each group separately, a linear and a positive relation was found for each group, respectively. The slope of the regression line for each group was significant. However, no differences were found between the different groups when the slopes were compared separately.

Regression analysis after the specimens had been regrouped according to the type of fracture showed that the regression line representing the type-1 fracture group did not differ from the regression line of the type-2 fracture group.

Discussion

When compressed axially the weakest part of the intervertebral joint is the end plate and the adjacent spongy bone (Hansson et al. 1980, Perey 1957, Plaue 1972, Rolander 1975). Recent studies have also shown that the end plate and the adjacent spongy bone is the weakest part of the intervertebral joint also during repeated submaximal loadings (Hansson et al. 1987).

The present study shows that the first part to fail during compression and simultaneous flexion, both within physiologic limits, was the end plate and the underlying spongy bone.

When the flexion was increased to 15°, the second type of failure occurred (type II). In comparison with the type I fracture, the type II fracture apparently involved other parts and components of the motion segment than type I. However, this was not reflected by a different relationship between the ultimate strength and the BMC in the heavily flexed specimens.

Because most lifting in vivo is performed while the lumbar spine is kept in a flexed position and because both disk pressure measurements and mathematical analysis have shown that flexion will cause the highest compressive load stresses in the vertebral bodies, it seems reasonable that the vertebrae in most people would adapt to these circumstances and resist more load in flexion (Granhed et al. 1987). Such a remodeling adaptation has already been found to reflect the changes in load sharing caused by different properties of the disk in the lumbar spine (Hansson et al. 1987). The absence of differences in the relationship between the BMC and the ultimate compressive strength during different degrees of flexion seemed to speak against such a remodeling adaptation owing to the higher load requirements on the vertebral motion segments during flexion

A similar correlation was found between the BMC and ultimate compressive strength irrespective of the degree of flexion used during compression. This indicated that the increased compressive stresses present during flexion of the lumbar spine had no significant influence on the remodeling of the vertebral bodies. Compression and flexion, the latter within normal ranges, caused a fracture pattern similar to that found earlier during axial loading and also during cyclic loading. Flexion at or beyond the upper limit of normal flexion caused a wedge fracture.

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