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From the Biomechanics Laboratory, Orthopaedic Department,
University of Göteborg, Göteborg, Sweden.

THE EFFECT OF ORIENTATION ON SOME MECHANICAL PROPERTIES OF FEMORAL CORTICAL SPECIMENS

By

CARL HIRSCH[1] and ODILIO DA SILVA[2]

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INTRODUCTION

Since Rauber in 1876 revealed different moduli of elasticity in longitudinal and transverse sections of cortical bone, attention has been paid to the way in which collagen fibers in bone are disposed. This information was a matter of controversy after the classical work of Gebhardt and Weindenreich had been published. They concluded that the fribrillar structures of human bone tissue was an arrangement of lamellae of alternating longitudinal and circumferential fibrils, organized in the longitudinal axis of the Haversian system. Recently *Smith* (1960) re-examined the subject and his findings were in conformity with their interpretation. *Maj & Toajari* (1937, 1938) in a series of papers, demonstrated that compact bone tissue of adults revealed three definitive axes of orientation, correlated with the cylindrical coordinates. They showed that the load required to fracture the specimen longitudinally to the long axis of bone was three times greater than that for specimens cut tangentially to the axis, and six times greater than that of specimens cut radially to the axis. The tangentially-cut specimens were about twice as strong as radially-cut specimens. They concluded that the resistance of compact bone to bending failure is proportional to the number of collagen fibers present in the plane of section of the bone, and that characteristic mechanical

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[1] Professor and Chairman, Department of Orthopaedic Surgery, Faculty of Medicine, University of Göteborg.

[2] Special Research Fellow of Conselho Nacional de Pesquisas, Rio de Janeiro, Brazil.

anisotropy of bone is the result of the distribution and direction of collagen fibers. *Dempster & Liddicoat* (1952) tested both wet and dry specimens and showed that cortical bone was significantly stronger longitudinally than it was in either the tangential or the radial direction, with no significant differences detectable between radial and tangential compression loads.

In this laboratory the mechanical properties of cortical bone have been reviewed recently (*Sedlin* 1965, *Sedlin & Hirsch* 1966) and an analysis of basic factors that control the physical properties was presented. It was intended to develop a model in which the behaviour of bone could be explained in mathematical terms.

Since it is generally agreed that the fiber arrangement is functionally important, the present study was undertaken to determine how the variability in direction of fibers in bone could influence its mechanical properties. Cortical specimens from human femurs were cut at 10 degrees increments. The first was made parallel to the longitudinal axis of the shaft of the femur and the last was cut at 90 degrees to that axis. The samples were subjected to bending tests and the slope, residual deformation, and energy dissipated, was recorded and calculated.

MATERIAL AND METHODS

This study was based upon 140 samples of human femoral diaphysis obtained at the time of routine post-mortem examination of 12 adult subjects. The ages ranged from 30 to 74 years, and both sexes were included. Many of the specimens were taken from patients who were at bed rest for varying periods of time prior to death.

The segments of femoral diaphysis were approximately 15 cm long and were taken 5 cm distal to the base of the lesser trochanter. After removal from the body they were prepared immediately or stored at —30° until such time as they could be prepared. The method of preparation and testing consisted of the following steps:

The medial and posterior surfaces of the diaphyseal specimens were ground under cold water using water-proof silicon carbide paper until these surfaces were perpendicular to each other.

Each specimen was then sectioned in its long axis using a band saw converted to the specifications outlined by *Bush* (1956) utilizing constant irrigation of the specimen with cold water. These new surfaces were ground until smooth.

The medial segment of the shaft was fixed, a protractor applied, and strips 10 mm broad were sectioned at intervals of 10 degrees in relation to the longitudinal axis of the shaft from 0 to 90 degrees. It was usually possible to obtain 10 strips at different angles from each specimen. These were labelled accordingly and placed in Ringer solution.

To establish whether there was any significant difference between the lateral and medial segments and the superior and inferior portions, all cases were compared. In 8 cases, two samples were cut from the lateral segment and compared

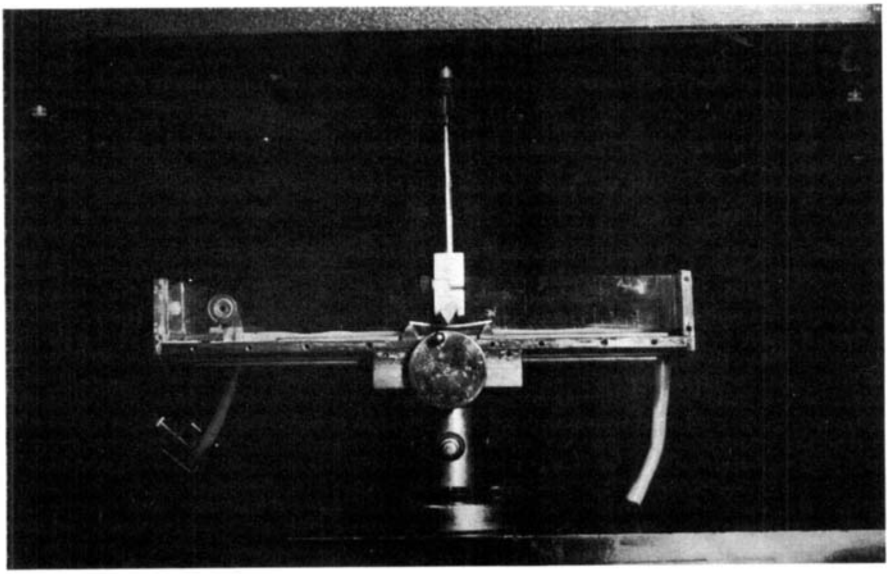

Figure 1. A specimen is being bent as an end supported center loaded beam.

with those taken from the medial segment. The first was cut at 90 degrees, from the proximal shaft, and the second was cut at 0 degree, from the distal. To evaluate the superior and inferior portion, a similar comparison was made in four cases. The lateral segment was sectioned superiorly into two strips, one at 90 degrees and other a 0 degree. The same was done in the inferior portion.

Reduction of the segments to test size was achieved with a special tool by grinding under constant cold water drip using water proof silicon carbide paper, as described by *Sedlin & Hirsch* (1966).

The size of the final specimens obtained was 20 mm, 2 mm and 1 mm. After preparation, the specimens were measured using a friction release micrometer accurate to 0.05 mm. All specimens with more than 1 per cent discrepancy in size were discarded or reworked. After measurements were completed, the specimens were placed in Ringer's solution and stored at —30° C until tested.

Tests were performed on an Instron TT–BM floor model tensile test machine, calibrated to an accuracy of ± 0.5 per cent.

Only bending tests were used in this study (Figure 1). The specimen was placed so as to become a centrally-loaded, simple support beam with a support length of 18 mm. It was progressively loaded and unloaded by means of a special loading head while submerged in Ringer's solution at room temperature (21° C ± 1). The test machine was programmed to cycle between zero load and 50 per cent full scale. Two tests were performed on each specimen with a two minute interval allowed for recovery. The position of the specimen was not changed between loads. The forces used were 0.100 and 0.250 kiloponds at a speed of 0.02 cm per minute.

The maximum deformation, the residual deformation and the energy dissipated in each cycle test was measured. The slope of each curve was calculated as the

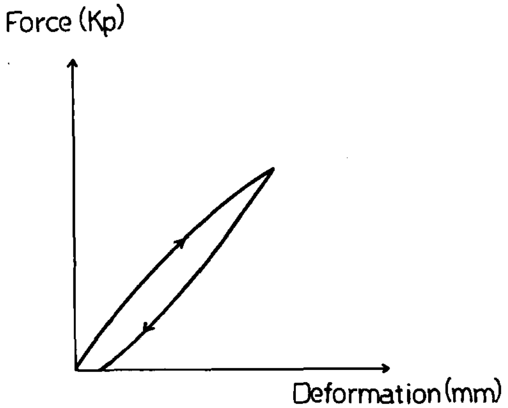

Figure 2. Load deflection curves produced by loading and unloading. The maximal deformation at the top of the curve and the residual deformation at zero stress in mm. The area between the curves represents the energy lost during the cycle of deformation.

quotient of maximum load (P) and maximum deformation (α) The area of the curves was measured by planimetry or geometrically.

Standard statistical methods were utilized in evaluating results.

ANALYSIS OF RESULTS

The central bending test gave different load-deflection curves for increasing and decreasing loads. These differences were correlated with the longitudinal axis of the shaft. In all tests a loop was described. These were not geometrically identical but the findings were quite similar.

During loading, the curve ascends while the deformation increases up to the maximum force applied. When unloaded, a descending curve is obtained as the bone tends to return to its previous form (Figure 2). The area between these curves represents the energy lost during the cycle of deformation. The distance between these lines at zero stress is the residual deformation. On re-testing the specimens after a two minute interval, using a different force, it was found that the relation between load and deformation was identical.

The results are summarized in Table 1, 2 and 3. There is a relationship between the physical properties of bone tissue and the orientation in which the bone strips were cut.

The figures in Table 1, where forces of 0.100 and 0.250 Kp were used, show that the stiffness of bone decreases progressively from 0 to 90 degrees. The differences between the values for the slope decrease progressively for each ten degrees. Figures 3 and 4 demonstrate this tendency to be a straight line relationship.

Table 1. Medial samples from femoral diaphysis slope (P/α).

Number of samples	Specimens cut at angle of	Average for 0.100 Kp	Standard error	Average for 0.250 Kp	Standard error
12	0°	174	6	179	5
7	10°	173	6	169	12
7	20°	154	10	165	9
12	30°	149	7	153	8
12	40°	134	5	136	4
11	50°	117	3	116	3
11	60°	110	5	109	6
12	70°	99	6	96	7
11	80°	94	7	88	8
12	90°	87	4	73	5

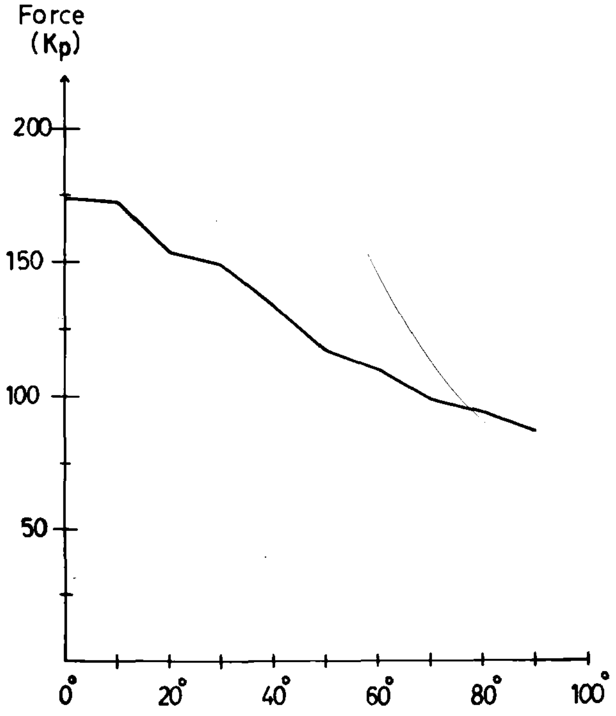

Figure 3. Diagram of slopes using 0.100 Kp force.

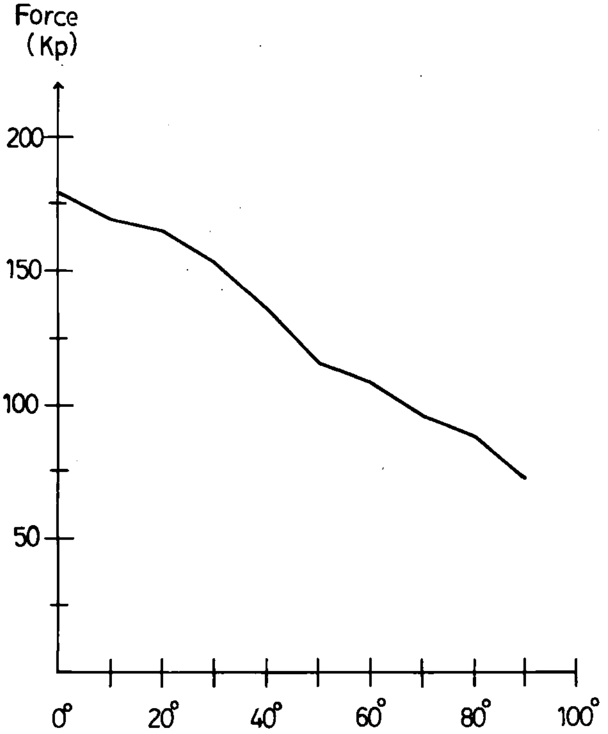

Figure 4. Diagram of slopes using 0.250 Kp force.

Table 2. Residual deformation (mm).

Number of samples	Samples cut at angle of	Average for 0.100 Kp	± Standard error	Average for 0.250 Kp	± Standard error
12	0°	0.04	0.005	0.02	0.005
7	10°	0.04	0.008	0.01	0.005
7	20°	0.03	0.007	0.02	0.005
12	30°	0.04	0.004	0.03	0.006
12	40°	0.04	0.004	0.02	0.007
11	50°	0.04	0.007	0.04	0.012
11	60°	0.03	0.004	0.06	0.019
12	70°	0.02	0.004	0.13	0.04
11	80°	0.03	0.008	0.20	0.09
12	90°	0.03	0.007	0.32	0.07

The residual deformation is expressed in Table 2. An applied force of 0.100 Kp gives a negligible residual deformation and a curve which is a straight line of variable obliquity for all sections. However, when 0.250 Kp is applied the shape of the curve is similar, with a small residual deformation up to 60 degress. Beyond this level, the stape of the curve shows increasing residual deformation and the entire curve becomes more horizontal, approaching the abcissa. The 90 degrees samples show this to extreme. In 3 instances the 80 and 90 degrees samples could not resist the maximum force applied and these specimens were unloaded at the point where the curve became horizontal to avoid failure of the specimens.

Table 3. Energy dissipated (Kp/mm).

Number of samples	Samples cut at angle of	Average for 0.100 Kp	± Standard error	Average for 0.250 Kp	± Standard error
12	0°	0.00018	0.00002	0.00029	0.00006
7	10°	0.00019	0.00004	0.00023	0.00011
7	20°	0.00013	0.00003	0.00032	0.00009
12	30°	0.00019	0.00002	0.00029	0.00007
12	40°	0.00018	0.00002	0.00021	0.00007
11	50°	0.00019	0.00003	0.00053	0.00024
11	60°	0.00015	0.00002	0.00136	0.00050
12	70°	0.00009	0.00002	0.00448	0.00188
11	80°	0.00015	0.00003	0.00704	0.00296
12	90°	0.00016	0.00007	0.01109	0.00246

The energy dissipated, Table 3, becomes greater as the residual deformation became greater.

Since this study was made on the medial segment of the femur it was important to establish if there were significant differences between the lateral and medial segments and between the superior and inferior sections. The evaluation of the test results by the Student's T test revealed that the slope, the residual deformation, and the energy dissipated of each curve was not significantly different from the medial segments (Table 4, 5 and 6). The same was true for the superior and inferior sections (Table 7, 8 and 9).

Table 4. Differences between medial and lateral segments slope (Pa).

Number of samples	Samples cut at angle of	Side	Average for 0.100 Kp	Student T test	Average for 0.250 Kp	Student T test
8	0°	Medial	173	not significant	185	
		Lateral	184		189	not significant
8	90°	Medial	85		67	
		Lateral	76	not significant	65	not significant

Table 5. Residual deformation (mm).

Number of samples	Samples cut at angle of	Side	Average for 0.100 Kp	Student T test	Average for 0.250 Kp	Student T test
8	0°	Medial	0.07	not significant	0.03	
		Lateral	0.07		0.02	not significant
8	90°	Medial	0.07	not significant	0.8	
		Lateral	0.03		1.0	not significant

Table 6. Energy dissipated (Kp/mm).

Number of samples	Samples cut at angle of	Side	Average for 0.100 Kp	Student T test	Average for 0.250 Kp	Student T test
8	0°	Medial	0.0002	not significant	0.0002	not significant
		Lateral	0.002		0.0002	
8	90°	Medial	0.002		0.01	
		Lateral	0.003	not significant	0.02	not significant

Table 7. Differences between superior and inferior section of the lateral segment of the femur slope (P/α).

Number of samples	Samples cut at angle of	Section	Average for 0.100 Kp	Student T test	Average for 0.250 Kp	Student T test
8	0°	Superior	176	not significant	191	not significant
		Inferior	184		191	
8	90°	Superior	83	not significant	72	not significant
		Inferior	100		91	

Table 8. Residual deformation (mm).

Number of samples	Samples cut at angle of	Section	Average for 0.100 Kp	Student T test	Average for 0.250 Kp	Student T test
8	0°	Superior	0.01	not significant	0.02	not significant
		Inferior	0.02		0.02	
8	90°	Superior	0.02	not significant	0.2	not significant
		Inferior	0.04		0.4	

Table 9. Energy dissipated (Kp/mm).

Number of samples	Samples cut at angle of	Section	Average for 0.100 Kp	Student T test	Average for 0.250 Kp	Student T test
8	0°	Superior	0.00006	not significant	0.0003	not significant
		Inferior	0.00014		0.0003	
8	90°	Superior	0.00012	not significant	0.01	not significant
		Inferior	0.00059		0.02	

SUMMARY

It is apparent from earlier investigations that the arrangement of collagen fibers play a basic role in the strength of bone tissue. The influence of small changes in direction of fibers on bone strength has not previously been evaluated. Our study demonstrates that the stiffness of cortical bone is closely related to the angles of which bone is cut being less stiff as the angle increases in relation to the longitudinal axis of the shaft. The residual deformation and the energy dissipated are proportional and vary with the angle.

RESUME

Des investigations antérieures ont fait ressortir que l'arrangement des fibres collagènes joue un rôle fondamental pour la force du tissu osseux. On n'a pas procédé antérieurement à une évaluation de l'influence que peuvent avoir de petits changements dans la direction des fibres sur la force de l'os. Notre étude démontre que la rigidité de l'os cortical est étroitement liée aux angles selon lesquels l'os est coupé. Il devient rigide au fur et à mesure de l'augmentation de l'angle. Par rapport à l'axe longitudinal du corps de l'os, la déformation résiduelle et l'énergie perdue sont proportionnelles à cet angle et varient avec lui.

ZUSAMMENFASSUNG

Aus früheren Untersuchungen geht hervor, dass die Anordnung von kollagenen Fasern eine wesentliche Rolle hinsichtlich der Stärke von Knochengewebe spielt. Der Einfluss von geringen Veränderungen in der Richtung der Fasern auf die Knochenstärke ist vorher nicht bestimt worden. Unsere Untersuchung erweist, dass die Steifheit kortikalen Knochens eng mit den Winkeln in Bezug auf die Längsachse des Schaftes unter denen der Knochen geschnitten wird, zusammenhängt, indem die Steifheit abnimmt wenn der Schneidungswinkel zunimmt. Die zurückbleibende Deformierung und die verschwendete Energie sind proportional und wechseln mit dem Winkel.

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