The Mechanical Properties of Bone[1]

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BONE is the material with which the orthopaedic surgeon deals. Consequently, some knowledge of its mechanical properties is of importance for an understanding of the mechanism and management of fractures, as well as the design of prosthetic or orthotic appliances and protective gear, *e.g.,* crash helmets. The behavior of a body under a load or force is a function not only of the form and structure of the body, but also of the mechanical properties of the material composing the body. For example, a steel beam will support a higher load before breaking and will behave differently under loading than will an oak beam of exactly the same shape and dimensions because of differences in the mechanical properties and structure of steel and of wood.

The mechanical properties of bone are determined by the same methods used in studying similar properties of metals, woods, and other structural materials. These methods are based on certain fundamental principles of mechanics, a knowledge of which is essential for understanding the terminology employed.

Mechanics, the science dealing with the effect of forces upon the form or the motion of bodies, has two subdivisions—statics and dynamics. *Statics* is the study of bodies at rest or in equilibrium as a result of the forces acting upon them. *Dynamics* is the study of moving bodies. The mechanical properties of materials are usually studied under static conditions, *i.e.,* under a slowly applied force or load, because the behavior of the test specimen can be more easily analyzed when the load is slowly applied.

A *force* is anything which tends to change the state of a body with respect to its motion or the relative position of the molecules composing the body. More simply stated, a force is a push or a pull. There are three primary kinds of forces: (1) *compressive* or pushing together forces, (2) *tensile* or pulling apart forces, and (3) *shearing,* or forces which make one part of the body slide with respect to an adjacent part (Fig. 1).

When a force is applied to a body, it produces stress and strain within the body. *Stress* (Fig. 1) is the ratio between the force and the area upon which it acts, *i.e.,* force per unit area. Stress is generally computed in terms of pounds per square inch (psi) or kilograms per square millimeter (ksm). Recently, some investigators of the strength characteristics of bone and other biological materials have been recording stress values in terms of kiloponds, dynes, or newtons per unit area, instead of pounds or kilograms because pounds and kilograms are units of mass as well as units of force. There will be no misunderstanding, however, if one specifies that stress values are in terms of *pounds force or kilograms force per unit area.* Stress is often used synonomously with strength, but the term has little value unless the kind of strength, *i.e.,* tensile, compressive, etc., is indicated. All strength values in the following discussion are in terms of *pounds force per square inch.*

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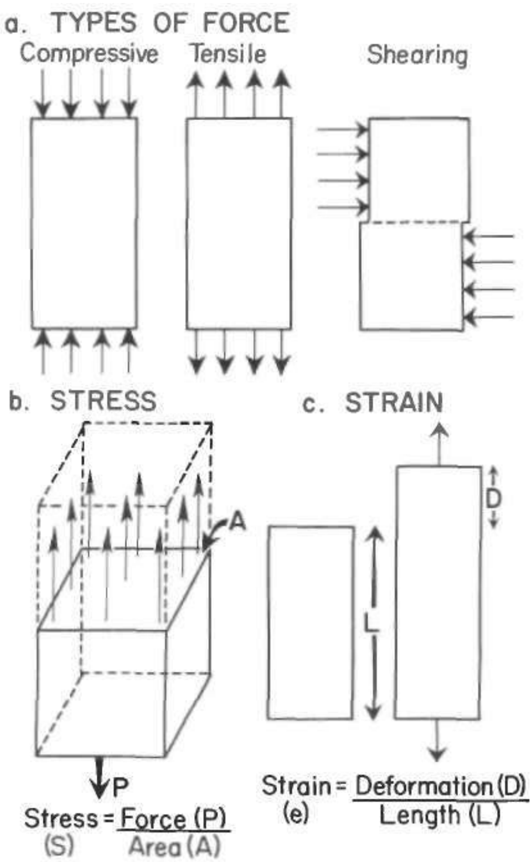

Fig. 1. Types of pure force—stress and strain.

Strain is a change in the linear dimensions of a body as the result of the application of a force (Fig. 1). Since there are no standard units of measurement for strain, it can be recorded as percentage, inches/inch, centimeters/centimeter, etc. Strain can be seen if it is sufficiently large, *e.g.*, as in stretching of a rubber band, but stress, which is only the ratio between force and area, is always invisible. The kind of stress and strain in a body is the same as the kind of force producing it.

When stress is plotted against strain, a *stress-strain curve* is obtained (Fig. 2). From a tangent drawn to the straightest part of the stress-strain curve the *modulus of elasticity* of the material, or the ratio between unit stress and unit strain, can be computed. The modulus of elasticity is a measure of the *stiffness* of a material, not its elasticity as one might assume from the name. *Elasticity* is the property of a material that allows it to return to its original dimensions after the removal of a force or load. The *energy* the specimen absorbs to failure can be determined by measuring the area below the stress-strain curve.

The method of choice in determining the tensile or compressive strength of a material is to make a test specimen of a standardized size and shape and test it under a pure tensile or a pure compressive force. Under these conditions the cross-sectional area of the specimen is known, or can be easily computed, and only one force—tension or compression—

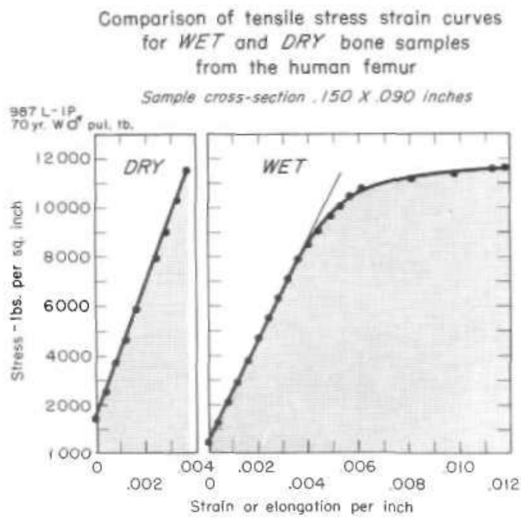

Fig. 2. Stress-strain curves for a dry- and a wet-tested specimen of compact bone from the posterior quadrant of the proximal third of the femoral shaft of a 70-year-old white man who died from pulmonary tuberculosis. The stress values are in *pounds force per square inch (7).*

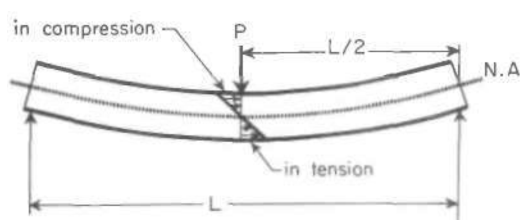

Fig. 3. Distribution of tensile and compressive forces in a body tested like a simple beam *(6). L* = length or span between supports; *N. A.* = neutral axis or plane; *P* = force or load.

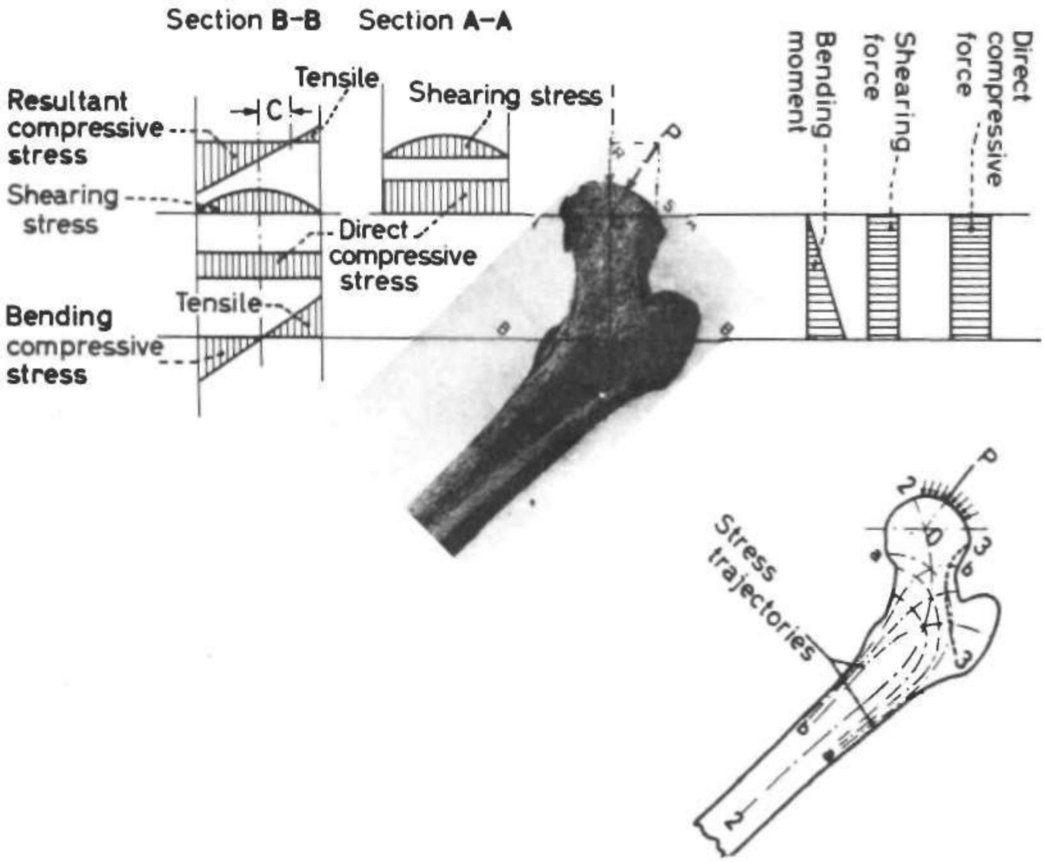

Fig. 4. Stress distribution in the neck of the femur *(20)*.

is involved. Furthermore, the force is uniformly distributed over the cross-sectional area of the specimen. Consequently, the ultimate tensile or compressive strength of the material can be easily calculated from the formula $S = P/A$, in which S is stress, P is force or load, and A is the cross-sectional area of the specimen (Fig. 1).

If the specimen is tested like a simple beam (i.e., supported at the ends and loaded midway between the supports) and bending occurs, tensile, compressive, and shearing forces are all involved. Tensile forces develop on the convex side of the bent specimen while compressive forces occur on the opposite (concave) side (Fig. 3). Both types of forces are maximum at the surface and decrease inwardly to zero at the neutral plane or axis. There are also shearing forces which, like the tensile and compressive forces, are not uniformly distributed over the cross section of the specimen. Under bending conditions, the force responsible for failure as well as its magnitude is more difficult to determine. The bending forces in the neck of the femur, as a result of the load applied to the head of the bone (Fig. 4), have been determined by Zarek *(20)*, an engineer who is currently working in biomechanics. For further discussion of forces in bending, see Harris' *Strength of Materials (11)*.

The speed at which a force is applied to a specimen influences the values obtained for some of its mechanical properties. McElhaney and Byars *(17)* found that the ultimate compressive strength and the modulus of elasticity of fresh and embalmed femoral cortical bone from cattle and man increased with higher strain rates of loading while the energy-absorbing capacity and the strain at failure decreased. The effect of high strain rates of loading on specimens of beef bone, cut and tested in different directions, has recently been investigated by Bird *et al.* *(1).*

Embalming also affects the mechanical properties of bone, at least those of compact bone. Thus, the mean ultimate tensile strength (in the long axis of the specimen and of the intact bone) is greater at the 0.01 significance level in embalmed wet- and dry-tested tibial specimens than in similarly tested unembalmed specimens *(4).* Furthermore, embalmed, wet-tested tibial specimens have a higher mean tensile strain, a greater mean single shearing strength (perpendicular to the long axis of the specimen) and are harder (Rockwell No.) than similarly tested embalmed specimens (5). However, the latter type of specimens has a higher mean modulus of elasticity. An analysis of variance showed that the increase in the hardness of the embalmed specimens was significant at the 0.01 level. As far as I am aware, there are no similar studies concerning the effect of embalming on the mechanical properties of spongy bone.

Two types or forms of bones are found in the foot—irregularly shaped bones (the tarsals) and miniature long bones (the metatarsals and the phalanges). The tarsal bones are essentially shells of compact bones filled with spongy bone, fat, marrow substance, blood, etc. The actual amount of osseous material in bones, such as the tarsals and the bodies of vertebrae, is not very great. According to Policard and Roche *(18)* the talus and the calcaneus are about 80 per cent nonosseous tissue. The percentage of bone in the bodies of 92 human lumbar vertebrae studied by Bromley *et al.* (2) varied from a maximum of approximately 24 per cent to a minimum of 15.5 per cent in males and from 21 per cent to 12 per cent in females at 5 and 70 years of age, respectively. As far as I am aware, there are no studies on the mechanical properties of spongy bone from the foot. Therefore, examination of such properties will be based on data obtained from the human femur.

Two types of specimens were used—a rectangular bar (the standard specimen) 0.79 cm. x 0.79 cm. x 2.5 cm. and a cube 0.79 cm. on a side. The specimens were obtained from the head, neck, greater trochanter, and condyles of the femur with the long axis of the standard specimens oriented in different directions.

The specimens were tested under direct compression in a Riehle 5000-lb. capacity testing machine, equipped with an automatic stress-strain recorder and calibrated to an accuracy of ±0.5 per cent. The low range scale of the machine (0-200 lbs.) was used with the load registered on the dial of the machine in units of 0.5 lbs. The specimens were loaded at a speed of 0.45 in. per min.

All specimens were tested wet to more nearly approximate the condition in the living foot. Drying of compact bone increases its ultimate tensile strength (in the long axis of the specimen), its modulus of elasticity, and its hardness (Rockwell No.) but decreases its single shearing strength (perpendicular to the long axis of the specimen) and its tensile strain (7, 8). Similar studies have not, to my knowledge, been made on spongy bone.

The ultimate compressive stress (strength) and strain, the modulus of elasticity, and the energy absorbed to failure were computed from stress-strain curves for wet-tested specimens. The density of air-dried specimens was determined with a strontium 90 densitometer developed by Evans, Coolbaugh, and Lebow *(9).* Dry specimens were used to avoid the effects of moisture that might be trapped within the interstices of the specimens. A total of 69 rectangular (standard) specimens and of 15 cubic specimens from 1 adult, white female, 3 adult, Negro males, and 6 adult, white males were tested. All specimens were kept in saline solution until tested. A minimum of 20 load-deformation readings were taken for each specimen during the test period.

The results of the study showed that the mean compressive stress (strength) of the cubic specimens was greater than that of

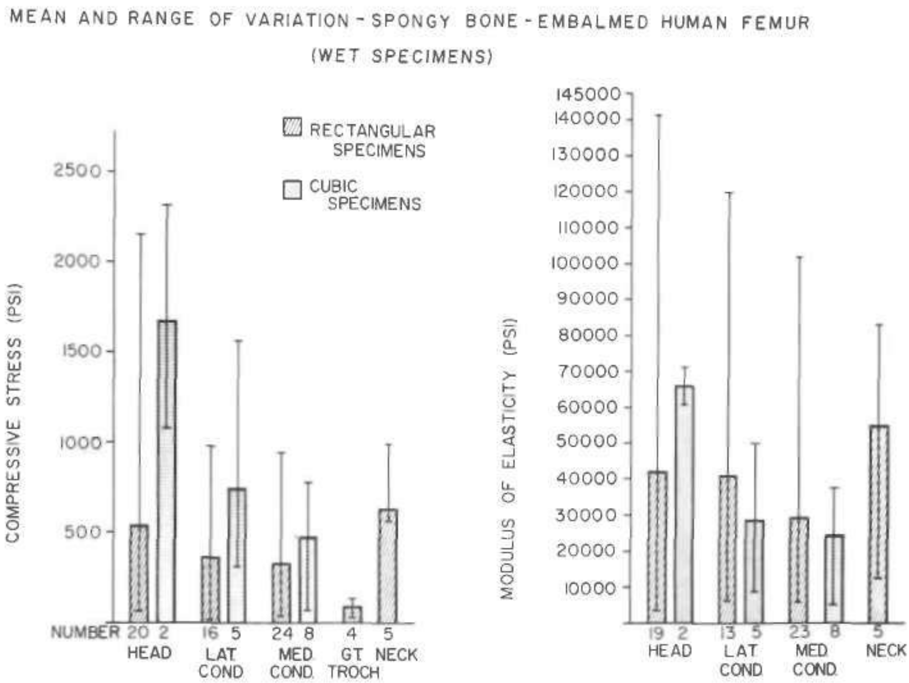

Fig. 5. Mean and range of variation in some mechanical properties of spongy bone from different regions of the femur. Compressive stress values in *pounds force per square inch*. *Gt. troch.* = greater trochanter; *Lat.* = lateral; *Med.* = medial; *Cond.* = condyle.

the rectangular (standard) specimens from the same region (Fig. 5). This phenomenon is characteristic of practically all materials. In cubic specimens high frictional forces developed between the ends of the specimen and the testing machine to resist the tendency of the specimen to be squeezed out of the machine. Furthermore, the upper part of the cube tends to be impacted into the lower part. Both of these factors contribute to higher values for compressive stress and modulus of elasticity in cubic than in specimens which are longer than wide. Because of these factors, it is felt that the values obtained from the rectangular (standard) specimens more accurately represent the true mechanical properties of spongy bone.

In the living body, most of the bones are subjected to bending action as a result of gravity, muscular activity during movement, and blows. Consequently, the bones are subjected to a combination of tension, compression, and shearing rather than to a single pure force. The question then arises as to why the strength of bone is usually determined by testing the specimens under a pure force. The answer to this question, on mechanical grounds, has already been given. There are, however, other valid reasons for testing the strength of bone under pure tension or compression.

Experimental studies with strain sensitive lacquers on bones within the living body as well as outside of it demonstrate that certain types of linear fractures of the skull, the pelvis, and the long bones all arise from failure of the bone from tensile stresses and strains produced in it by bending (3). The determination of the tensile strength of bone under pure tension thus has direct application to the mechanics of fractures of those types. Clinical experience also indicates that tensile forces are important in the production of many types of fractures.

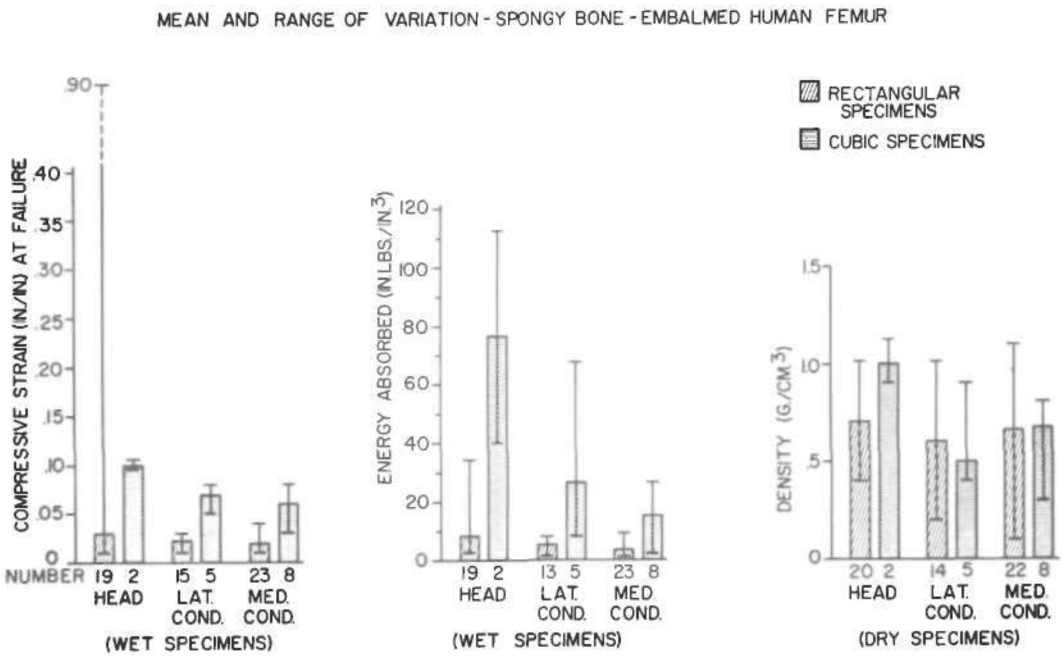

Fig. 6. Mean and range of variation of some mechanical properties of spongy bone from various regions of the femur.

Compression fractures are quite common in the bodies of the vertebrae, especially those in the lumbar region, and in the calcaneus, the most frequently fractured of the tarsal bones *(12)*. Compression fractures of the talus also occur. There is, consequently, a sound practical reason for investigating the compressive strength of the tarsal bones, especially the calcaneus and the talus although, to my knowledge, it has not been done. The rationale for determining the strength of spongy bone from the femoral head and condyles under direct compression is that these regions of the bone are normally subjected to compression forces in the erect posture *(13)*. Specimens from other regions were similarly tested for comparative purposes.

When the results of the tests were compared according to the region of the bone from which the specimens were obtained, without regard to the direction of loading, several differences were found. The rectangular (standard) specimens from the neck had the highest and those from the greater trochanter the lowest mean compressive stress. Among the cubic specimens the highest and the lowest mean compressive stresses were found in specimens from the head and the medial condyle, respectively.

Regional variation was also found in the modulus of elasticity (stiffness) of the specimens (Fig. 5). The mean stiffness of the rectangular specimens exceeded that of the cubic specimens from the same region except for the specimens from the head. The rectangular specimens from the neck and the medial condyle, respectively, had the highest and the lowest mean modulus. The maximum and the minimum stiffness means of the cubic specimens were found in those from the head and the medial condyle, respectively.

Comparison of the mean compressive strain, mean energy absorbed to failure, and mean density of the rectangular and cubic specimens from different parts of the femur also reveals interesting differences (Fig. 6). The cubic specimens showed somewhat more variation in the mean compressive strain than did the rectangular ones, the strain being greatest in the specimens from the head and least in those from the medial condyle. Little difference was found in the mean compressive strain of the rectangular specimens, those from the head having a slightly greater strain than those from the condyles. The cubic and the rectangular specimens from the head had the highest while those from the medial condyle had the lowest mean energy absorbed to failure. However, the former specimens showed more regional difference than did the latter. The mean density for both types of specimens was greatest in those from the head and least in the ones from the lateral condyle.

A statistical analysis of the above data from the rectangular (standard) specimens revealed the following significant differences between the means. The mean compressive stress of the strongest specimens (from the neck) was greater, at the 0.02 significance level, than that of the weakest specimens (from the greater trochanter). The difference between the mean compressive strain of the specimens from the head, which had the highest, and that of specimens from the medial condyle, which had the lowest, was significant at the 0.01 level.

The mean energy absorbed by the specimens from the head was significantly greater, at the 0.02 level, than that absorbed by specimens from the medial condyle. The differences between the means for the other mechanical properties of the rectangular specimens were not statistically significant. The number of cubic specimens tested was not sufficiently large for statistical analysis.

Comparison of the maximum compressive stress and modulus of elasticity (Fig. 7) of the rectangular and cubic specimens

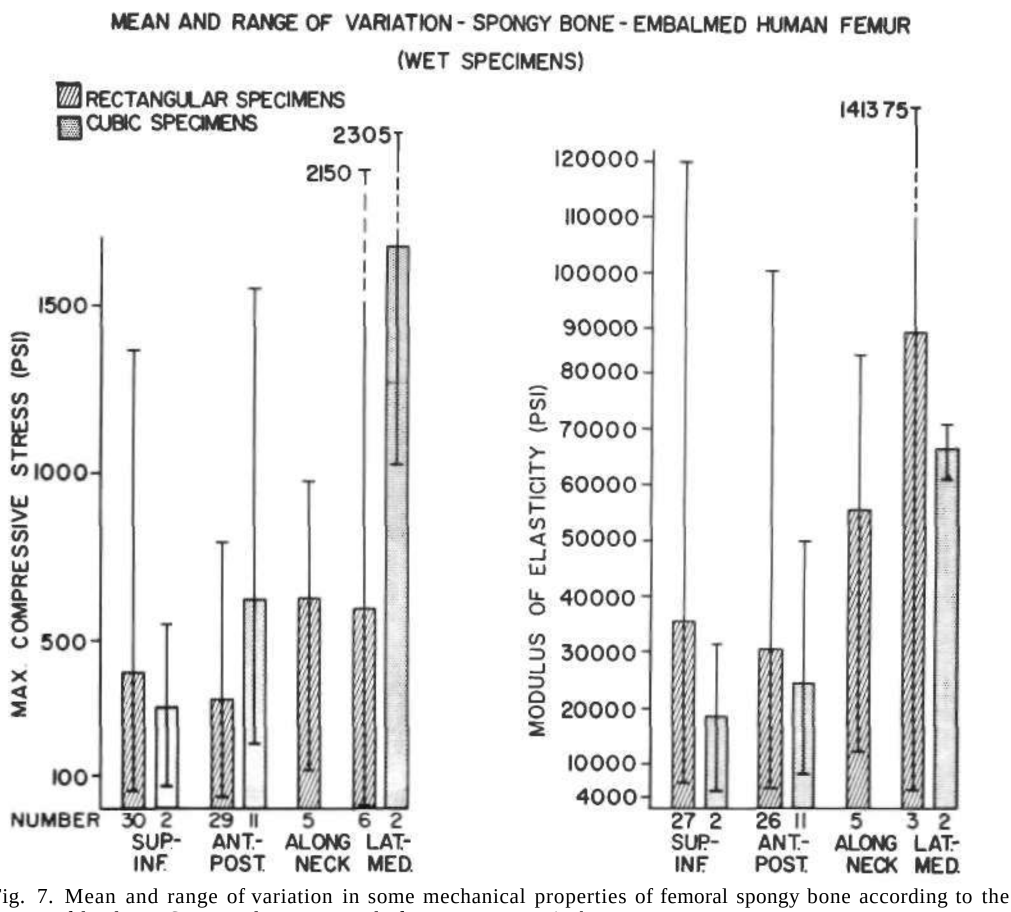

Fig. 7. Mean and range of variation in some mechanical properties of femoral spongy bone according to the direction of loading. Stress values in *pounds force per square inch.*

according to the direction of loading showed that spongy bone is an anisotropic material, *i.e.*, a material that is not equally strong in all directions. The rectangular specimens loaded in the direction of the long axis of the neck of the femur showed the highest, while those loaded in the anterior-posterior direction showed the lowest mean compressive stress. Among the cubic specimens, the highest mean compressive stress was found in specimens loaded in a lateral-medial direction and the lowest in specimens loaded in a superior-inferior direction.

The rectangular specimens loaded in a lateral-medial direction had the highest mean modulus of elasticity and those loaded in the anterior-posterior direction the lowest. The cubic specimens loaded in a lateral-medial direction had the highest mean modulus of elasticity while the lowest was found in the specimens loaded in a superior-inferior direction.

Considerable variation was also found in the energy absorbed to failure, the compressive strain at failure, and the density of the specimens when evaluated with respect to different directions of loading (Fig. 8). The rectangular specimens loaded in a lateral-medial direction had the highest mean energy-absorbing capacity whereas those located in an anterior-posterior direction had the lowest. The highest mean energy-absorbing capacity among the cubic specimens was found in those loaded in a lateral-medial direction and the lowest in the specimens loaded in a superior-inferior direction.

The rectangular specimens loaded in a lateral-medial direction had the highest average compressive strain and those loaded in the direction of the long axis of the neck had the least. The compressive strain of the cubic specimens loaded in a lateral-medial direction far exceeded that of all other specimens. The lowest com-

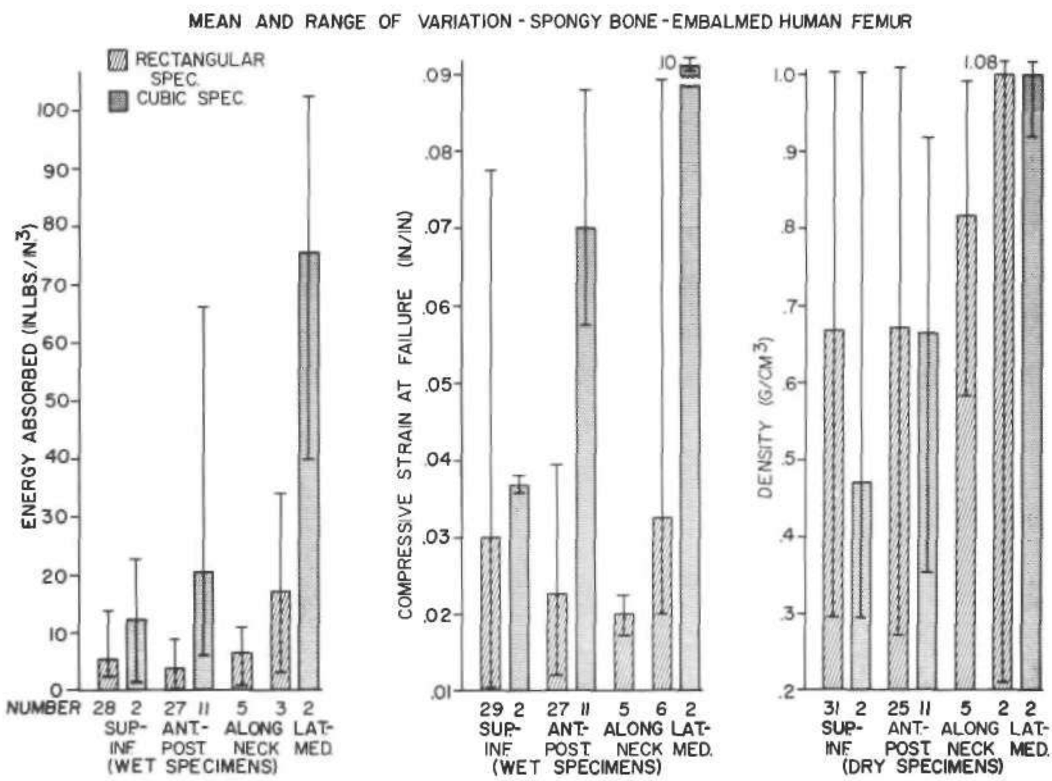

Fig. 8. Mean and range of variation in some mechanical properties of femoral spongy bone according to the direction of loading.

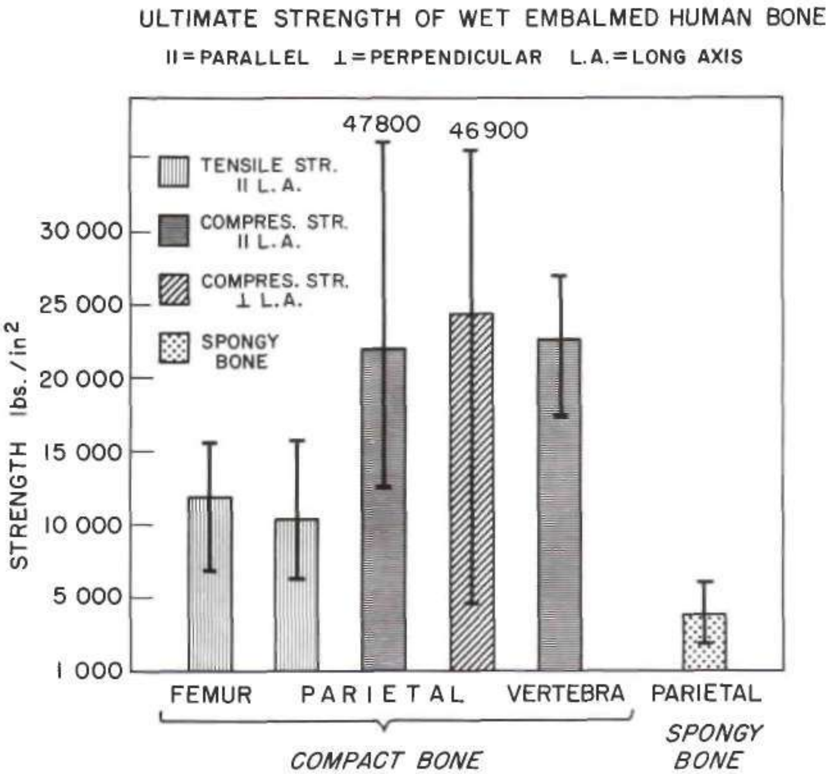

Fig. 9. Mean and range of variation in strength of various bones according to type (compact or spongy) and. direction of loading (6).

pressive strain among cubic specimens was found in those loaded in the superior-inferior direction.

Surprising differences were found in the density of specimens cut in different directions. The density of rectangular and cubic specimens cut in the lateral-medial direction was the same but greater than that of any other specimens. The rectangular specimens cut in the superior-inferior and in the anterior-posterior direction were the least dense. Cubic specimens were the least dense when cut in the superior-inferior direction. These differences in density of the specimens suggest directional variation in the orientation and abundance of trabeculae in various parts of the femur.

A statistical analysis of the means for the various mechanical properties with respect to the direction of loading revealed the following significant differences. The variation between the energy absorbed by rectangular specimens, loaded in the lateral-medial direction, was significantly greater at the 0.01 level than that of the specimens subjected to anterior-posterior and to superior-inferior loading. The difference between the maximum compressive strain (found in lateral-medial loading) and the minimum strain (found in specimens loaded in the direction of the long axis of the neck) was significant at approximately the 0.04 level. No other significant differences were found between the means for the other mechanical properties when analyzed with respect to the direction in which the specimens were cut and loaded.

Although spongy bone is much weaker than compact bone (Fig. 9), its foam-like structure makes it a good energy-absorbing material, as demonstrated experimentally more than a century ago by Dr. Physick *(19)* and more recently suggested by Evans, Pedersen, and Lissner *(10).* The presence of fat, marrow substance, and blood in the

TABLE 1. RANGE OF VARIATION IN THE FATIGUE LIFE OF HUMAN METATARSALS TESTED WITH A LOAD OF 15 LB. *(15)*

Metatarsal	Dry-Tested		Wet-Tested	
	No. of Specimens	Repetitions to Failure	No. of Specimens	Repetitions to Failure
II	7	2,000- 1,343,000	3	151,000-11,117,000
III	9	1,000- 870,000	2	6,860,000-13,908,000
IV	6	14,000- 2,273,000	3	150,000- 1,177,000
V	8	10,000-10,297,000	2	195,000- 521,000

interstices of spongy bone in the living condition enhances its energy-absorbing capacity by making it act like a quasi-hydrostatic system. The capacity of bone to absorb energy is one of its important mechanical properties as far as fracture mechanics is concerned because, as pointed out by Lissner and Evans *(16),* all physical injuries arise from the absorption of energy. Most fractures are produced by impacts or blows and thus involve energy absorption.

Another mechanical property of bone to be considered is its fatigue life. This is especially important in relation to march, stress, or fatigue fractures which are most common in the metatarsal bones although they have also been reported in other bones. These fractures are thought to be the result of repetitive loading such as occurs during marching, hence the name "march" fracture.

The only investigation known to me on the fatigue life of intact bones is one we made several years ago *(15).* In this study the strength of intact human metatarsal bones was determined by loading them to failure in a Sonntag Flexure Fatigue machine equipped with an automatic counter (which recorded the number of cycles to failure) and shutoff. The chief advantage in using this type of fatigue machine is that it has an inertia force-compensator spring which absorbs or eliminates all unknown inertia forces. Consequently, the force in the specimen being tested, regardless of its rigidity, is equal to the known force produced by the oscillator assembly.

Forty-one bones were tested with a force of 15 lbs. (the maximum that could be applied with our machine), 3 bones with 12 lbs., and 8 bones with 10 lbs. Only the second through fifth metatarsals were tested because the first one was too large for the fatigue machine. The influence of moisture upon the fatigue life of the specimens was investigated in 10 bones by allowing water to drip on them during a test. The bones were not degreased and all were tested at room temperature. None of the bones exhibited any known pathologic condition. In order to hold the bone in the fatigue machine during a test, the ends were embedded in Selectron 5026 plastic. The number of repetitions to failure was automatically recorded and the machine shut off as soon as the specimen broke. A cycle means the bone is bent once up and once down.

Comparison of the results obtained for the wet- and the dry-tested specimens showed that drying tended to decrease the fatigue life of the bones (Table 1). The probable explanation is that drying increased the modulus of elasticity of the bone and hence the specimens were stiffer. The number of repetitions to failure, with a 15-lb. force, varied from 1,000 to 10,297,000 for the dry specimens and from 150,000 to 13,908,000 for the wet specimens. Metatarsals 2 and 3 showed the greatest fatigue life when tested wet. No consistent relations were found between the fatigue life of the bones and their size or age of the individuals from whom they were obtained. The type of fracture produced experimentally (Fig. 10) was similar to some reported *(14)* in the clinic literature (Fig. 11).

It is interesting to speculate how long an individual must walk before the metatarsals would be subjected to the same num-

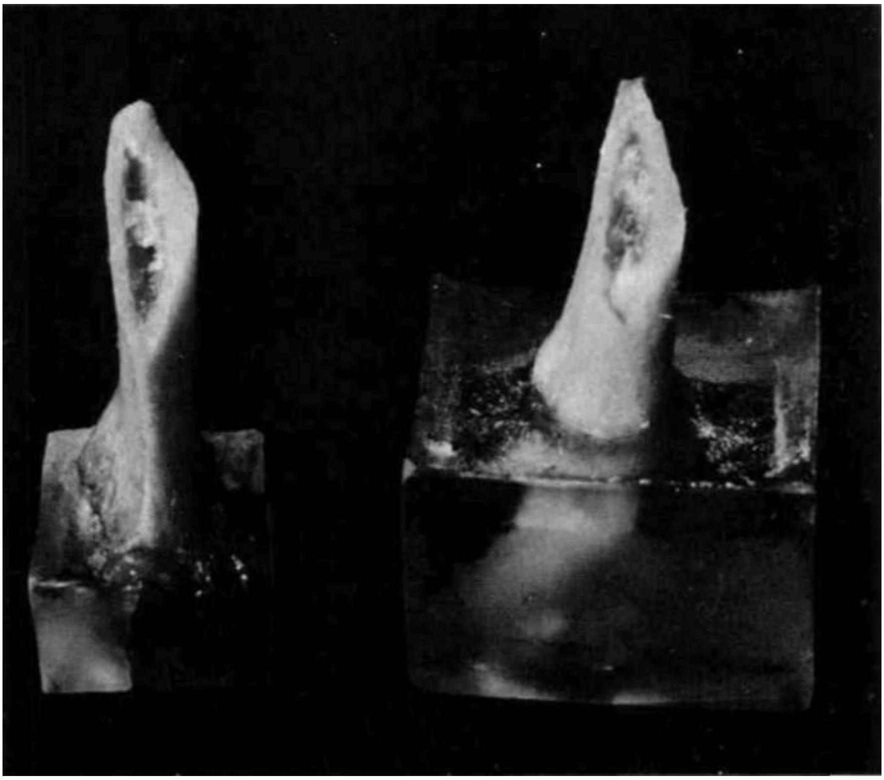

Fig. 10. Experimentally produced fatigue fracture of an intact human metatarsal bone.

Fig. 11. A clinical fatigue fracture of a metatarsal bone *(14).*

ber of repetitions at which failure occurred in our experiments. If it were assumed that an individual walked at the army pace of 120 steps per min., walking 50 min., resting 10 min., one would have to walk continuously for almost a month before the second metatarsal would be subjected to the number of repetitions at which the failure occurred in the present study. During each cycle of loading, the bone was bent up and down in a vertical plane. The fracture was probably a tensile failure initiated on the side which, at the instance of failure, was the convex or tensile side.

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