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Review Article

The Mechanical Properties of Cortical Bone*

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From the literature concerning the mechanical properties of bone tissue, data can be extracted to support or contradict many hypotheses, not because of any inherent ambiguities in the data but rather because of the absence of standardization or unification of experimental methods, approaches, and goals. The inclusion of all published experiments on the mechanical properties of bone in this review would most likely leave every one discouraged or baffled. Since we have the advantage of hindsight, the older literature may be culled for those investigations whose conclusions and findings remain valid and useful in the light of recent studies in which refined methods were used. To make these investigations readily available we have included them in a supplementary bibliography (Appendix).

The first limitation established for this review was that only those investigations which treated bone as a material and not as an anatomical structure would be considered. We therefore shall discuss mainly the testing of bone specimens cut from the cortex of bones and not the testing of whole bones. The testing of whole bones certainly gives important information, and there are questions which can be answered only by testing whole bones, but if the analyses of such tests consider anything more than rigid idealizations of bone architecture, then a characterization of the material is essential. Currey [12] has discussed the interrelation of these two experimental approaches in posing and answering biological questions.

We are going to discuss those mechanical properties which we consider of prime significance, namely the elastic constants (relating stress to strain), the viscoelastic parameters (relating stress, strain, and time), the plastic parameters (describing permanent deformation), and the strength or ultimate properties (stating conditions of fracture or failure). The concepts of fatigue failure and crack initiation will also be discussed.

The question of the relationship between the size of the specimen and the objective of the test performed must be considered because to some degree bone is a structure at all levels of organization including the molecular level, and this complex nature makes the size of the machined specimen an important variable to be considered in all tests. Because of the heterogeneity of bone, a working definition of the "material" under test is needed. The majority of investigators in the past have used specimens with a cross-sectional area in the range of from four to twenty square millimeters, which is a satisfactory size range since specimens with these dimensions contain at least several haversian systems. We also used this cross-sectional area to define bone tissue. Thus we did not examine the properties of single osteons but rather the average properties of multiple osteons (or lamellae). Use of such a specimen, however, gives no information concerning the mechanical properties of bone tissue at the level of the haversian systems, capillary networks, and cells. Ascenzi and Bonucci [2,3], in order to determine the tensile and compressive properties of secondary osteons, used single secondary osteons as specimens of bone tissue, while other investigators, notably Katz [23], considered bone tissue at an even lower level, building a model of a composite material to test the mechanical properties of the organic and inor-

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ganic constituents of bone. The study of the properties of secondary osteons necessitates characterization of primary osteons and of surface or circumferential bone lamellae as well, since all three are present in the cortex of long bones. Also, the combination and interaction of the different types of bone must be considered in order to develop a model of cortical bone as a whole. Katz's approach not only needs precise characterization of the properties of the constituents themselves, but also requires knowledge of the integration of the constituents into the material called cortical bone on the molecular, macromolecular, fibrillar, lamellar, and osteonal levels. Such properties are not known.

The majority of investigators have therefore chosen to investigate a "material" which is one level below that of whole bone. This is also an approach used successfully by engineers for complex materials if the anisotropy (properties differing in different directions within the material) is not too great (a change of less than one order of magnitude in measured characteristics with a change in direction). Models have been derived with this approach for woods, plywood, soils, rocks, crystals, some reinforced materials and, as we shall see, bone.

The second limitation on the information reviewed was the treatment of the cortical bone specimen during fabrication, storage, and testing. Sedlin and Hirsch [37] considered a large number of variables in the preparation and testing of cortical bone and showed that embalming and drying are important variables. It is not known if embalming affects the capability of bone tissue to deform plastically or if this effect is uniform in relation to specimen orientation. Investigators using embalmed wet bone, however, have never noted any remarkable amount of plastic deformation and Tsuda [42] found a decrease in the amount of deflection necessary to produce failure in binding tests on embalmed bone compared with similar tests on fresh bone. This difference may be due to fixation of the organic portion of the tissue by the embalming solution and a subsequent change in mechanical characteristics. Therefore the results of experiments which used embalmed bone are not included in this review. The data on the effect of embalming on the physical properties are equivocal [15,31,37], and if fresh or frozen bone [36] is available it is certainly preferable.

The loss of plastic deformation before fracture in dry bone and the importance of this loss as a factor in the determinations of mechanical properties has been shown by Burstein and co-workers [7] in fresh bone and by Evans [15] in embalmed bone. In this connection it should also be noted that bone *in vivo* is not dry and while its precise state of hydration is not known, its condition probably is closer to that present in a wet than in a dry test. Rewetted dry bone has been used, but drying can induce cracks [17] which change the stress response and strength characteristics. Hence, in this review we shall for the most part consider bone specimens which were kept wet during all stages of preparation and testing. The reader is referred to the Appendix for information relative to bone tested under other conditions.

Elastic Properties of Bone Tissue

The relationships between induced stresses (force per unit area) and the resulting strains (change in length or angle) for a particular material are expressed as proportionality constants which are termed elastic constants. The elastic properties of any given material may differ according to the direction in which testing is performed. If the properties are different in every direction the material is said to be anisotropic and has thirty-six elastic constants. If, however, there is one plane in which the elastic properties are the same in every direction on that plane, the material is said to be transversely isotropic and has only five elastic constants. Finally, if there is no directional dependence of the elastic properties (complete symmetry) the material is called isotropic and has only two elastic constants, which are referred to in engineering terminology as Young's modulus and Poisson's ratio. Young's modulus (E) or stiffness is the slope of a stress-strain curve obtained for uniaxial stress, or in other words, the ratio between stress and strain. Poisson's ratio (ν), which is defined as the negative of the ratio of transverse strain to longitudinal strain in the direction of uniaxial loading, is a measure of the material's ability to conserve volume when loaded in one direction. For example, in the usual response of an object its sides expand under a compressive load and contract under a tensile load. This response is Poisson's effect.

Another elastic constant is the shear modulus (G), defined as the ratio of induced shear to the resulting shear strain. In the isotropic case this constant is dependent on Young's modulus and Poisson's ratio and is quite often determined from a torsion test. The shear modulus relates the angular distortion (shear strain) to the shear stress in the material. For further discussion of these material constants and their mathematical manipulation, the reader is referred to the works of Lekhnitskii [28] and Love [29].

Young's Modulus

This is the first of the two independent constants for an isotropic material and is the ratio of normal stress to normal strain produced in a simple one-dimensional tension or compression test. Some of the values for Young's modulus are given in Table I, where E is the modulus in a direction perpendicular to the long axis of the bone and E' is the modulus in a direction parallel to the long axis.

The inherent structural symmetry of a material, be it layered plywood, crystals, fiber-reinforced composites, or bone, can often be used to predict the elastic symmetry. Employing this fact Lang [27] determined five constants for bovine phalanx using a transversely isotropic matrix. From measurements of the transit velocities for ultrasonic waves he computed the elastic constants which would give these transit times using the five-constant model (Table I). At present, the transversely isotropic model seems best since bone histologically displays the symmetry of a transversely isotropic material.

TABLE I

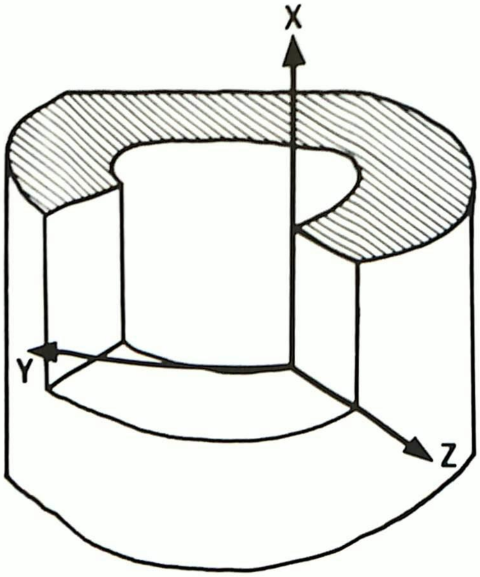
YOUNG'S MODULI

Bone Species and Authors	Type of Loading	Constants*† ($\times 10^9$ N/m)	Comments
Human			
Dempster and Liddicoat, 1952	Compression, very low strain rate	E' = 8.69 E = 4.19 transverse E = 3.76 radial	Small cubes of rewetted dry femur, tibia, humerus; strain measurement derived from crosshead movement
	Compression, very low strain rate	E' = 14.1	Cylinders of rewetted dry femur, tibia, humerus; strain measurement technique not reported
Bovine			
Sweeney and assoc., 1965	Tension, low rate	E' = 17.2 E = 9.2	Femur, histology not reported; strain measurement derived from crosshead movement
	Compression, low rate	E' = 16.5 E = 9.9	
Lang, 1970	Ultrasonic	E' = 22.0 E = 11.3	Phalanx, histology not reported
Burstein and assoc., 1972	Tension, strain rate: 0.1 sec^{-1}	E' = 17.2 ± 5.10 E = 11.1 ± 1.77	Femur, plexiform; strain measurement with an extensometer

* E' = Young's modulus in direction parallel to the long axis of the bone. E = Young's modulus in direction perpendicular (transverse or radial) to the long axis of the bone.

† By way of comparison, E of methylmethacrylate (plexiglass) is 8.6; of Douglas fir (68 per cent moisture), 13.4; and of steel, 210.

Dempster and Liddicoat [14] also determined the elastic moduli in three directions (longitudinal, transverse, and radial) for small cubes of human bone and found no statistically significant difference for the elastic moduli in the radial and the transverse directions (within the plane of isotropy of the five-constant model), but the modulus in the longitudinal direction was approximately twice as large (Fig. 1). However, these workers used rewetted dry bone and apparently mixed specimens from femora, tibiae, and humeri together. Furthermore, their compression tests were done on small cubes (approximately four to five centimeters), a circumstance that raises the question of edge effects in their specimens. When specimens are in the form of small cubes it is difficult to obtain uniform stresses on the faces and edges of the cube which are in contact with the surfaces of the testing machine. If these surfaces are not flat and parallel the stresses are not uniform. Furthermore, as the specimen is compressed its sides must move outward in a direction perpendicular to the direction of the load (Poisson's effect). As the load increases, however, friction limits this movement at the sites of specimen-machine contact



X – LONGITUDINAL
Y – TRANSVERSE
Z – RADIAL

FIG. 1

The longitudinal direction (x) is perpendicular to the isotropic plane (y-z). The z direction corresponds to the radial direction, while y is the transverse direction.

and the specimen under load becomes barrel-shaped. This so-called edge effect is accentuated when the test specimens are small and hence it must have been a significant factor in the results of Dempster and Liddicoat. Still another problem in these tests was the use of the travel of the platens of the testing machine to measure strain without reporting any correction for the stiffness of the machine. The difficulty in evaluating these authors' data is shown by the two values they obtained for Young's modulus in the longitudinal direction (8.69×10^9 N/m^2 and 14.1×10^9 N/m^2), when the only change was in the dimensions of the specimen.

Sweeney and co-workers [41] used machined, standardized tension and compression specimens prepared from pieces of cortical bone cut in the two perpendicular directions (longitudinal and transverse) for the development of stress-strain curves. These investigators, however, calculated strain by measuring the distance traveled by their testing machine crosshead and used an unreported, but implied, low strain rate. The problems involved in comparing material constants determined at different strain rates will be discussed under "Viscoelastic Behavior."

The final entry in Table I, our own [6], gives values for Young's moduli which were calculated using strain measured by a clip-on strain-gauge extensometer and are for a strain rate (0.1 sec^{-1}) within "physiological" limits. The use of an extensometer on the gauge length of the test specimens obviates most errors in the gripping arrangement and the need for corrections for testing-machine stiffness.

Table II shows some of the values found in the literature for Young's modulus (E') for bone tissue considering only one loading direction. All of these data are for experiments which used machined specimens of bone tissue and a uniaxial stress field oriented along the long axis of the bone. We have expressly not included in this listing values obtained using whole bones. Obtaining values for Young's modulus from testing bone usually entails a bending test with the compounded problem of a complex cross section and structure, both of which must be accounted for in normalization of the results. Hence, values derived from whole-bone tests are of little use in discussing material characteristics but may be used for comparative studies, as mentioned by Currey [11]. Bending tests have also been interpreted with the assumption that the material behaves in a linearly elastic fashion and that strain can be calculated from midpoint deflection of the beam. These assumptions hold only if the stress in the material is kept below the yield level or the initial slope of the bending moment-deflection curve is used. Sedlin and Hirsch [37] reported values for Young's modulus (E') which differed by a factor of approximately two when the only change in the experimental conditions was the test of configuration, that is, tension versus bending. This difference indicates that they may not have fulfilled the requirements (tension stress below the yield stress) to be valid, and hence introduced errors. If the bending test is carried beyond the yield point an elastic-plastic analysis must be performed as described by Burstein and co-workers [7]. The errors produced by the use of an elastic assumption after the yield point can be at least as great as the value determined. While the values for Young's moduli for longitudinal and transverse directions obtained in bending tests may be used for purposes of comparison with the above reservations, the values for elastic constants obtained from specimens excised from whole bone at angles between the extremes of the longitudinal and transverse directions are of limited use unless an anisotropic beam theory is used in reduction of the data.

The experimental methods used in the determination of the values for Young's modulus shown in Table II were not all satisfactory. Some of the drawbacks associated with the values are listed in the column headed "Comments." A most difficult problem is the measurement of actual strain (elongation or shortening) in the gauge section of the specimen rather than the travel of the grips or testing machine platens. We believe that the most acceptable figures in the table for Young's modulus are those of McElhaney and co-workers [31], of Simkin and Robin [39] in tension only, and of our laboratory [6] for human and beef bone, since for these determinations strain was measured by a strain-gauge extensometer clipped to the reduced section of the specimen. Ko [25] attempted to minimize errors in strain determination by optically measuring the changing distance between two wires attached to the test section of the specimen. No strain rate was given for this work, but the technique implies a very slow "unphysiological" rate. Other investigators have attached strain gauges to the specimen surface, but this necessitates drying of the bone surface and hence yields values of questionable biological significance. It is interesting that only Sweeney and co-workers [41] found Young's moduli to be equal in both tension and compression. For small deformations there are no theoretical reasons for Young's modulus to be different in the two loading configurations. Comparison of the values for the modulus in tension and compression obtained by McElhaney and associates [31] is impossible, since for their compression tests these authors used a long thin specimen (prone to buckling behavior) which was supported laterally. If, however, the moduli for similar strain rates obtained by McElhaney and co-workers [31] in tension and by McElhaney and Byars [30] in compression are compared, a ratio of approximately three to four is found. The

TABLE II

YOUNG'S MODULUS OF WET BONE
(LOADING DIRECTION PARALLEL TO BONE AXIS)

Bone Species and Authors	Type of Loading	Young's Modulus ($\times 10^9$ N/m²)	Comments
Human			
Dempster and Liddicoat, 1952	Tension, low strain rate	14.1	Dry rewetted femur, tibia, humerus mixed; extensometer used
Kimura, 1952	Compression, low strain rate	10.4	Femur; movement of testing-machine head determined by dial indicator for strain measurement
Ko, 1953	Tension, low strain rate	17.3	Femur; strain measurement by optical lighting (see text)
Sedlin, 1965	Bending, unknown strain	15.8	Femur; testing-machine travel used for strain
Sedlin and Hirsch, 1966	Bending, no strain rate given	15.5	Femur, mid-portion; deflection of mid-point in 3-pt. bending used for strain; crosshead displacement used to determine strain in tension test
	Tension, no strain rate given	6.0	
Abendschein and Hyatt, 1970	Ultrasonic	24.5	Femur and tibia
Burstein and assoc., 1972	Tension, strain rate: 0.1 sec^{-1}	14.1	Femur; extensometer used for strain
Bovine			
Kimura, 1952	Tension, low strain rate	24.5	Femur, unreported histology; strain measurement same as Ko
	Compression, low strain rate	8.5	Femur, unreported histology; crosshead movement measured by dial indicator
McElhaney, 1965	Tension, strain rate: 3 sec^{-1}	20.5	Femur, unreported histology; strain measured from distance between platens
	Compression of cubes Strain rate: 0.1 sec^{-1} Strain rate: 1.0 sec^{-1}	 24.1 27.6	Femur, unreported histology, strain measured from distance between platens
Sweeney and assoc., 1965	Tension, no strain	17.2	Femur, unreported histology, stored in 50 per cent alcohol before machining
	Compression, no strain rate	16.5	Strain measured from crosshead motion
Lang, 1970	Ultrasonic	22.0	Phalanx, unreported histology
Burstein and assoc., 1972	Tension, strain rate: 0.1 sec^{-1}	24.5 ± 5.10	Femur; extensometer used
Simkin and Robin, 1973	Tension, low strain rate	23.8 ± 2.21	Tibia, unreported histology; extensometer used
	Compression, low strain rate	7.1 ± 1.05	Tibia; crosshead motion used to measure strain; strain rates same for tension and compression; tests took as long as five minutes

values for the modulus in compression reported by Simkin and Robin [39] and by Kimura [24] are unexpectedly and inexplicably low and may have been influenced by the compliance of their testing machines.

Poisson's Ratio

The second independent constant for an isotropic material, Poisson's ratio, is the ratio of strain in a direction perpendicular to that of a uniaxial stress to the strain in the

TABLE III

POISSON'S RATIO

Bone Species and Authors	Type of Loading	Value*†	Comments
Bovine			
Lang, 1970	Ultrasonic determination	$\nu' = 0.482$ $\nu = 0.397$	Phalanx, unknown histological type
McElhaney, 1965	Compression, strain rate: 0.1 sec^{-1}	$\nu = 0.28$	Femur, unknown histological type. Dried
Human			
Ko, 1953	Tension, low rates	$\nu = 0.08$ to 0.45	Femur; ν varied with load level; strain measurement techniques unknown

*ν' = axially loaded longitudinal specimen; ratio of strain in transverse direction to strain in longitudinal direction. ν = axially loaded transverse specimen; ratio of strain in radial direction to strain in transverse direction.

† By way of comparison, ν for methylemethacrylate (plexiglass) is 0.4; for Douglas fir (68 per cent moisture), 13.4 (ν' being 0.27); and for steel, 0.2.

direction of uniaxial stress. The reported values for these constants are shown in Table III. Here again comparison of the values is quite impossible due to the different variables including species, moisture, and strain rates. The value attributed to Ko was taken from Swanson's [40] review article, which states that the ratio in fact varied as the load level was changed and reached the lower value just prior to fracture of the specimen. The method used for the strain measurements in this study was not described. McElhaney and Byars [30] used a strain gauge glued to the specimen surface for strain measurement and therefore dried the surface of their bovine bones. Lang [27], as mentioned previously, used an ultrasonic technique with wet bovine bone at the upper extreme of physiological strain rate. No values for Poisson's ratio for human bone tissue deformed at rates which approach physiological conditions have yet been published.

Shear Modulus

The shear modulus, which is the ratio of induced shear stress to the resulting shear strain, for the isotropic case was determined by Ko [25] for human bone tissue (Table IV), using a slow-rate torsion test. The independent shear modulus of the transversely isotropic model was found ultrasonically by Lang [27] for bovine phalanx but the histological type of the bone was not specified.

Experimental Variables

The influence of the anatomical site of provenance of the specimen on mechanical properties was investigated by both Sedlin [36] and Evans and Lebow [18]. They found no statistically significant differences in the elastic moduli in bone tissue obtained from four anatomical quadrants (anterior, medial, posterior, and lateral) of the bones tested. Sedlin's bone material was fresh whereas Evans and Lebow's was embalmed. Hirsch and

TABLE IV

SHEAR MODULUS

Investigator	Material	Shear Modulus ($\times 10^9$ N/m²)	Comments
Ko, 1953	Human	0.31	Torsion test, low strain rate
Lang, 1970	Bovine phalanx	0.54	Ultrasonic test, unreported histology

daSilva [22], using only the initial slope from bending tests done on fresh human femoral samples, also found no difference in elastic modulus for different anatomical sites from the same bone.

Other important variables for which data on elastic constants obtained by sufficiently precise methods are not available include: the histological type of bone (especially important for bovine bone in which the histological type depends on animal age), age, mineral content, species, and the bone from which the specimen was obtained. Currey [11] looked at the question of mineralization for the case of whole bone in bending, and Ascenzi and Bonucci [2,3] considered the case of individual secondary osteons. Both investigations showed that the stiffness increases as the degree of calcification increased. Ascenzi and Bonucci found no difference in the tensile mechanical properties of individual osteons from an old and a young individual.

Viscoelastic Behavior of Bone Tissue

If the characteristic constants of a material are affected by the *rate* of deformation, then the material is said to be viscoelastic. All biological materials display at least some viscoelasticity. Strain-rate dependency of the constants for bone have been investigated by McElhaney and Byars [30], Sedlin [36], Burstein and Frankel [5], Currey [10], and Sammarco and co-workers [35]. Sedlin developed a qualitative rheological model which accounted for the elastic, viscoelastic, and plastic elements of the mechanical behavior of bone-tissue specimens, but none of the constants for this model were presented. Sedlin's preliminary work on tissue handling and specimen preparation have already been mentioned. McElhaney and Byars investigated the response of cubes of bovine bone to compression loads at various strain rates and found that Young's modulus increased, ultimate stress increased, and strain to failure decreased with increasing strain rate (Table V). Currey studied deformation and recovery of dry and wet bone specimens loaded as cantilever beams for long periods of time (weeks). With this creep test, he found that increased deformation with time was essentially recoverable. The investigations of Burstein and Frankel and of Sammarco and associates were concerned with the response of whole bones to torsional loads applied at different strain rates, and therefore can only be used in a qualitative manner in speaking of the behavior of bone tissue. Their findings were in agreement with those of McElhaney and Byars for increasing strain rate and ultimate strength of the bones, but in contrast they found that deformation to fracture increased with increasing strain rate. The fact that they used whole bones and different species from those used by McElhaney and Byars makes any comparison of their results difficult. Hence, while the viscoelastic behavior has been well documented for bone tissue, good data on the quantitative aspects of this behavior at physiological strain rates, especially for human tissue, are lacking.

It is also possible to determine the elastic and viscoelastic properties of bone with the specimen used as a spring. By attaching a mass to one end of a piece of bone and forcing

TABLE V*

Strain Rate† (1/sec)	Ultimate Compressive Stress ($\times 10^6$ N/m^2)	Young's Modulus ($\times 10^9$ N/m^2)	Maximum Strain (*Per cent*)
0.001	176	18.6	1.88
0.01	207	20.0	1.82
0.1	231	24.1	1.75
1.0	252	27.6	1.25
300.00	283	33.1	1.00
1500.00	365	42.1	0.90

* For fresh bovine bone of unknown histological type. Table adapted from McElhaney and Byars, 1965.
† Strain has no units (for example, mm/mm) so that strain per unit time (strain rate) has the units 1/second.

vibratory motion at the free end, the elastic and viscoelastic properties may be calculated from the motions of the mass and of the vibration source. Laird and Kingsbury [26] performed such experiments on wet bovine bone using a vibration source in the four-octave range from one to sixteen kHz. They found that the elastic and viscoelastic properties remained approximately constant at each frequency but these values were somewhat higher than those obtained quasistatistically at slow rates of loading. The elastic moduli obtained by these workers were in the ratio of 1.6:1.3:1.0 for the longitudinal, transverse, and radial directions. It appears, therefore, that the viscous effects of bone tissue do not grossly (that is, by an order of magnitude) alter the response of bone to loading at different frequencies.

Plastic Deformation

Plastic deformation is the deformation from which a loaded specimen, on release of the load, does not recover. Our laboratory [9] showed that bovine specimens, cut with their long axis parallel to the bone axis and subjected to tensile loads, exhibited considerable plastic deformation. A typical stress-strain recording showing this deformation is reproduced in Figure 2. Fully 70 per cent of the total deformation to failure was shown to be plastic. Plastic deformation was demonstrated or postulated by other investigators. Sedlin [36] published a very similar stress-strain diagram for a tension test of human bone but stated that the behavior of a Hookean (linearly elastic) body is a close approximation of the bone's behavior. This, of course, was true only up to the point of initiation of plastic strain. Dempster and Liddicoat [14] showed a stress-strain curve for wet human compact bone and stated that the curve deviated slightly from the straight line before the bone specimen fractured. We [6] showed a plastic component of the strain in human bone similar to that in bovine bone and therefore suggested that Dempster and Liddicoat were not obtaining the full strain capability of the tissue. This same suggestion applies to the work of Sweeney and associates [41] and of McElhaney and co-workers [31], who showed ultimate

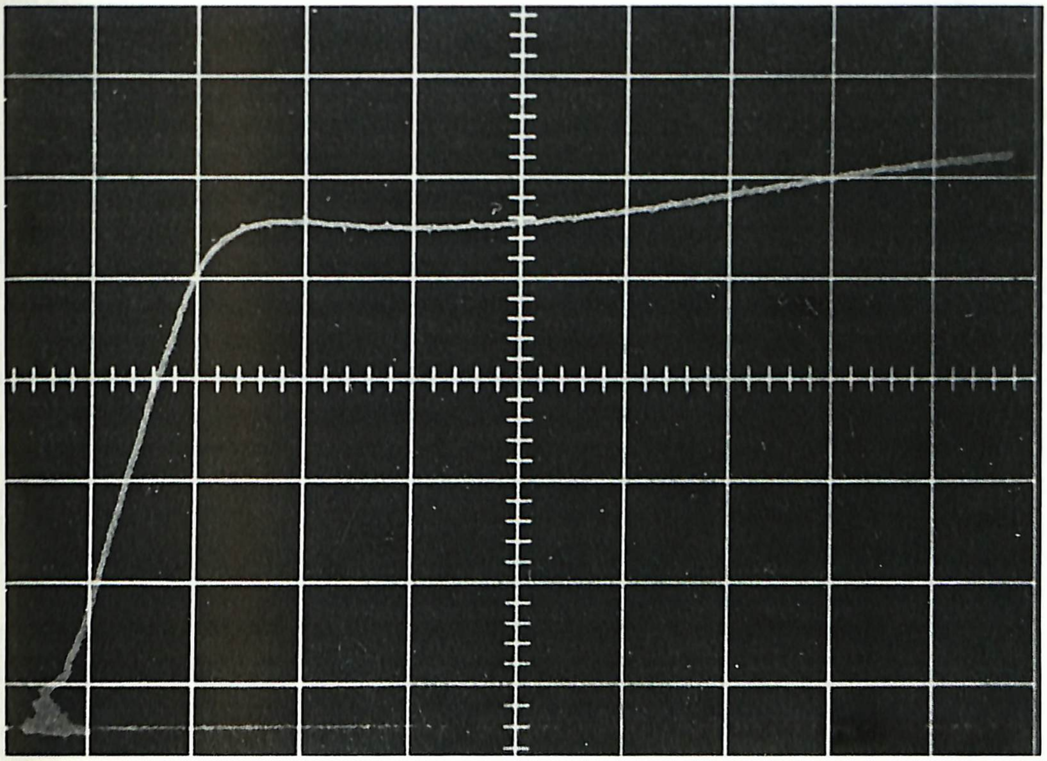


FIG. 2

Load-deflection curve from a longitudinal tension test. (Load on vertical axis, measured in newtons [n], and deformation on horizontal axis, measured in centimeters.) Initially, the load and deformation are linearly related and fully recoverable (elastic range). After the curve deviates from the initial straight portion, the plastic phase is entered and recovery from strain is no longer complete.

tensile strains of 0.010 and 0.012, respectively. The ultimate tensile strains reported by us [6] for a strain rate of 0.1 sec^{-1} (10 per cent increase in length per second) were 0.029* for bovine bone and 0.46 for human bone. Currey [10,11] also described and discussed the plastic deformation of bone and its biological significance.

The amount of plastic deformation exhibited by a bone specimen depends very much on the care with which the experimenters keep the specimen wet throughout its preparation and testing. When kept suitably wet and loaded along the axis of the bone, the loading curve obtained resembles that of a material that undergoes work-hardening showing a positive (upward) slope on the stress-strain record after yield (Fig. 2). Thus, if the bone material is handled in a way that does not allow the work-hardening or plastic-deformation behavior to be achieved, the nature of the ultimate properties is changed. The energy absorption of a material which is ductile (shows plastic deformation) is greater than that of one which is brittle but has a similar ultimate stress. This enhanced energy absorption of wet bone was discussed by Evans and Lebow [18] for embalmed material

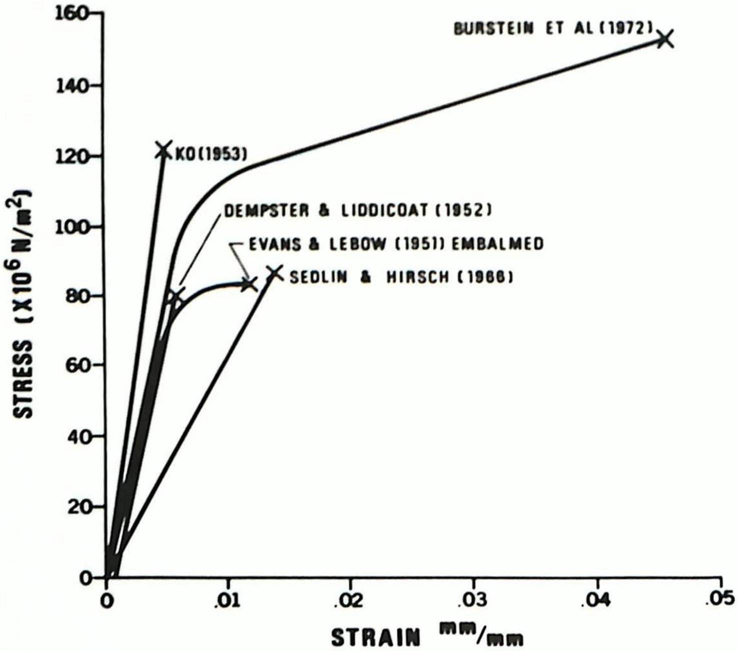


FIG. 3

Stress-strain curves for compact human bone constructed using values for Young's modulus, strain to failure, and ultimate stress reported by the authors noted. Stress is expressed as units of force (newtons) per unit area (square meters) while strain is deformation (millimeters) per unit length (millimeters).

which showed only 0.016 millimeter per millimeter as the maximum strain to failure. We [6] showed that if human bone tissue *in vivo* did display such stress-strain behavior, the energy absorption before fracture in tension would be increased by a factor of seven. (In Figure 3 compare the areas under the stress-strain curves which represent the work or energy necessary for fracture). The direction in which the load is applied to the specimen also affects plastic deformation. If the direction of the tensile load is perpendicular to the long axis of the bone, the material shows little or no plastic deformation [6].

The mechanism responsible for plastic tensile strain in bone is certainly not the same as that causing plastic strain in metals, that is, shear flow, or a gross sliding of metal along oblique planes within the specimen. Such flows are accompanied by gross shifting of atomic bonds along slip lines. We [6] postulated that a mechanism of "pull out" or "void formation" may be responsible for bone's ductile behavior in tension, based on the observation that light transmission in a bone specimen is decreased after the yield point is

* Strain is the ratio of change in length to original length. Since it is the ratio of two similarly dimensional quantities it has no units of its own. Thus 0.029 represents a 2.9 per cent change in length.

reached and the specimen enters the plastic phase. By "pull out" we mean a separation of material or a breaking of adjacent molecular bonds on parallel longitudinal planes. When compressive loads are applied, bone exhibits plastic deformation due to another mechanism. Hence, we speak of a structural plasticity, or plastic deformation which depends on the geometric displacement of materials within the structure rather than the ductility of the constituents themselves. Chamay [8] described "prefailure slip lines" in the cortex on the compression side of dog ulnae repeatedly loaded in a buckling configuration. He attributed these slip lines to a shear failure of the bone but noted that they were absent on the tension side of the bone (Fig. 4). Elucidation of the mechanism by which bone exhibits plastic deformation will probably encompass both these findings.

Strength

The bulk of the literature describing the mechanical properties of bone concerns itself with the determination of intrinsic strength. Usually this is defined as the maximum stress sustained by the bone specimen without fracture in some loading configuration, usually tensile, compressive, bending, torsion, or direct shear. How the maximum stress is to be measured dictates quite often the shape of the specimen used.

Testing Configurations

Tensile tests conventionally are performed on specimens fabricated with a gauge section reduced in size (so as to lessen the influence of the device used to grip the specimen) and are subjected to a uniaxial tensile load. The ultimate stress is found by dividing the highest load sustained by the original cross-sectional area. (This technique is based on the assumption that the amount of reduction in area during testing will be at least one order of magnitude less than the original area.) Axiality of the load is ensured by suspension of the specimen between knife edges or universal joints.

Compressive tests are a little more difficult to perform, since the edge effect from the testing machine and the axial inaccuracies are harder to eliminate. The ultimate compressive stress (or strength) is calculated using the same method and assumption regarding area change that are used for the tensile tests.

Bending tests are often used to determine tensile strength of bone but they must be interpreted from an elastic-plastic viewpoint if they are to be used to obtain absolute values, not simply for comparison. In other words, if the elastic beam formulation:

$$\sigma = \frac{Mc}{I}$$

where σ = stress in extreme fiber; M = bending moment supported by the beam; c = distance from the neutral axis to extreme fiber; and I = moment of inertia of the cross section is used, the calculation of the tensile stress in the extreme fiber becomes less accurate as the beam material displays more plastic deformation. With a material which displays the amount of plastic deformation found in bone in our laboratory [6,7] calculation of the ultimate strength using a bending test and assuming bone to be elastic will give values which overestimate the stress in the extreme fiber by 50 to 100 per cent [7]. Since most of the bending tests done on bone specimens prior to 1972 used an analysis designed for elastic materials, in discussing work done on the tensile strength of bone we shall separate those values determined by a bending test from those utilizing a uniaxial tension test.

Loading for the determination of the shear strength of a material may be applied to a specimen by the use of a torsion test, short-beam shear test, or a single or double shear test (Fig. 5). Each of these testing configurations has its problems.

A short beam can be used to subject material to short stresses, but unfortunately the test also induces tension and compression stresses since some bending occurs. The shear stresses, which are highest in the center of the cross section, are lower in magnitude than

FIG. 4

Scanning electron micrograph of surface of a specimen subjected to longitudinal compression (load direction vertical) showing presence of shear cracks, making an angle of approximately 40 degrees to the vertical.

the tension and compression stresses (which are highest at the outer regions of the cross section). Therefore, unless a material has a shear strength that is much lower than its tension or compression strength, failure will not occur in shear and the shear strength will not be measured. Both single shear tests (scissor action) and double shear tests (a type of punch test) also introduce many support and edge artefacts. Pure shear was applied to a specimen in a torsion test but the resulting spiral fracture of the bone specimen was attributed to a "tension weakness" by Dempster and Liddicoat [14] and to a failure in tension by Sweeney and co-workers [41]. Recently, however, we [6] found that the spiral fracture actually has a shear initiation and therefore the torsionally induced fracture is a measure of shear strength. (See section on Fracture Initiation.)

Ultimate Strength of Bone in Different Testing Configurations

The values for ultimate strength reported in the literature are shown in Tables VI through IX. As with the elastic constants, comparison of these values is difficult considering the many variables involved in the testing of bone-specimen strength. Wall and associates [45] studied many of the variables and suggested how they should be standardized. One of the considerations is the bone and site selected for testing. Differences have been shown in the strength of bone tissue from different bones. Using embalmed wet bone, Evans [17] found that specimens from the human femur are 14 per cent weaker in tension than specimens from the tibia and fibula, the specimens from the tibia and fibula having the same strength. Ko [25] showed differences in the tensile strength of specimens from different bones, the sequence from strongest to weakest being: radius, fibula, tibia, humerus, and femur. According to him the tensile strength of the tibia was approximately

17 per cent higher than that of the femur. Almost the reverse sequence for compression strength was reported by Yokoo [46], the femur being strongest, followed by the tibia (the strength of the tibia being 98 per cent that of the femur), the fibula, humerus, radius, and ulna (ulnar strength being 77 per cent of that of the femur). Therefore, magnitude and significance of the strength differences vary not only with the type of test and loading conditions, but also with the bone from which the specimen was obtained, and any discussion of the strength of bone tissue should probably be restricted to the same bone.

Differences in strength of bone specimens taken from different sites of the same bone have been reported by some investigators. These differences, however, have been small, and a consistent sequence of site strengths has not been established. Using a tension test, Ko [25] found that specimens taken from the medial quadrant of a femur are stronger than those from other quadrants. Evans and Lebow [18], using the same testing configuration but on wet embalmed material, found that in the femur the lateral quadrant produced the strongest specimens and the anterior quadrant, the weakest. Sedlin [36] found no difference in the bending strengths of fresh femoral specimens from the anterior, lateral, and medial quadrants, but specimens from the posterior quadrant were significantly weaker. Differences were also found in the strengths of specimens taken at different points along the length of the femur. It would therefore seem wise either to compare bone strength from some specific site or to take samples from each quadrant at some chosen level, as for example the mid-portion of the diaphysis, and to consider the mean.

Ultimate Strength and Histological Characteristics

The effect of the histological type of fresh bovine bone on the ultimate tensile stress was investigated by Currey [9]. He found a strong negative correlation between the number of haversian systems and tensile strength, with the strength of totally haversian bone being about 30 per cent less than that of laminar bone. He attributed this decrease in strength to the combined effect of decreased bone (secondary osteons have a larger central canal), and the lower degree of calcification of the secondary osteons. Evans [16], working with embalmed human tissue, similarly found that a large number of osteons and osteonal fragments in the specimen reduced the ultimate tensile strength. Aoji and associates [1] reported that for bovine bone the ultimate compressive strength was inversely proportional to the amount of interstitial lamellae in the cortex.

Ultimate Strength and Strain Rate

As was the case in the determination of the elastic constants, the viscoelastic nature of bone makes the strain rate used in any test an important variable (Table V). Most investigators did not report the strain rate used in their tests. The rate of travel of the testing machine crosshead alone does not define the strain rate without further information on specimen gauge length and total strain to fracture. Good data on fresh human compact bone, with careful consideration of strength with strain rate as a variable, are not available.

Ultimate Strength and Anisotropy

The question of variation of strength due to specimen orientation was approached by several authors. A difference in tensile strength (Table VI) was shown in the longitudinal (along the bone axis) direction, as opposed to the transverse direction. The large difference shown by Dempster and Coleman [13] for the strength in these two directions was perhaps due to the fact that these workers used rewetted dry bone. Evans [17] found that after drying there were cracks in cortical bone along the cement lines between osteons. If these cracks caused stress concentrations, strength determined for dry bone tissue would show a considerable decrease. Such a stress-concentration effect would be marked when the tensile stresses were applied perpendicular to the plane of the crack; that is, in the transverse direction. Dempster and Coleman's ratio of longitudinal to transverse tensile

TABLE VI

TENSILE STRENGTH

Bone Species and Authors	Ultimate Stress ($\times 10^6$ N/m^2)	Load Direction	Comments
Human			
Dempster and Liddicoat, 1952	78.8	Longitudinal	Femur, tibia and humerus mixed, dry rewetted; low strain rate
Ko, 1953	122 ± 1.1	Longitudinal	Femur; low strain rate
	140 ± 1.2	Longitudinal	Tibia; low strain rate
	146 ± 1.5	Longitudinal	Fibula; low strain rate
Dempster and Coleman, 1960	95.3 ± 27.0	Longitudinal	Tibia, dry rewetted; no strain rate
	9.9 ± 2.9	Transverse	
Melick and Miller, 1966	138	Longitudinal	Tibia, less than 60 years old
	119	Longitudinal	Tibia, more than 60 years old; very low strain rate, 2 minutes to fracture
Sedlin and Hirsch, 1966	86.5	Longitudinal	Femur; no strain rate
Burstein and associates, 1973	151 ± 18	Longitudinal	Femur; strain rate: 0.1 sec^{-1}
Bovine			
Currey, 1959	112	Longitudinal	Femur, plexiform; very low strain rate, 2 minutes to fracture
Sweeney and assoc., 1965	129	Longitudinal	Femur stored in alchohol, histology not reported; no strain rate
		Transverse	
Burstein and assoc., 1972	172 ± 22.0	Longitudinal	Femur, plexiform; strain rate: 0.1 sec^{-1}
	52 ± 8.0	Transverse	

strength for human bone would probably have approached that found by Sweeney and associates [41] and by us [6] for bovine bone, had their specimens not been allowed to dry.

The anisotropic strength behavior and the differences in total strain to failure were explained by Maj and Toajari [32] as being due to the number of collagen fibers oriented in the direction of the load. While these authors' specimens were in the shape of a plate and they reported only the loads necessary for fracture, they found a pronounced strength anisotropy in bovine bone with loads applied in the longitudinal, transverse, and radial directions, the strengths being in the proportions of 6:2:1 respectively. Evans and Vincentelli [20] also considered the ''predominant'' collagen-fiber orientation in relation to mechanical properties of embalmed human bone. In these two investigations the direction of the collagen fibers was determined by means of a polarizing microscope, a method which has recently been challenged by Boyde and Hobdell [4]. Using the scanning electron microscope they found that the collagen fibers are in a ''predominant'' direction only in very small domains within the lamella, and that only a few micrometers away, in another domain, the fibers are in some other ''predominant'' direction. Other evidence to relate bone-tissue strength to collagen-fiber direction is lacking. Hence it is difficult to draw any conclusions.

Ultimate Strength and Age

Some investigators showed a dependence of bone strength on the age of the individual. Melick and Miller [33] found that the tensile strength of the adult human femoral cortex declined approximately 4 per cent per decade when tested with a low strain rate that produced fracture in two minutes. Such strain rates, of course, are well below those produced by trauma. The mean values for specimens from patients less than and more than sixty years old are shown in Table VI. A similar rate of decay with age was found by Hazama [21] for shear strength, that of sixty to eighty-year-old subjects being 85 per cent that of twenty to thirty-nine-year-old subjects. In contrast to these data, Evans and

Lebow [18], using embalmed femoral samples, found no decrease of tensile strength with age, as did Sedlin and Hirsch [37] using a bending test on fresh femoral samples. The work of Vose and co-workers [44] on embalmed tissue from senile osteoporotic femora showed that the bending strength was higher in these specimens than in those from the normal population. The question of the age-dependent strength characteristics of human bone tissue is therefore by no means answered.

Ultimate Strength and Ash Content

Melick and Miller [33] tested the tensile properties of fresh human femoral bone specimens and found no correlation between the ultimate strength and ash content of the specimens. However, other investigators observed that intrinsic strength increased with increasing ash weight or content. Ascenzi and Bonucci [2,3] found 20 per cent greater tensile strength and 70 per cent greater compressive strength in fully calcified single osteons compared with those only initially calcified. Vose and co-workers [44], testing embalmed human bone specimens, showed an increase in the bending strength with an increase in the ash content. On the structural level, Currey [11] and Vose and Kubala [43] showed a similar increase in bending strength as ash content increased.

Ultimate Strength and Surface Wetness

We have already discussed the question of testing wet versus dry bone and presumed that the wet condition is closer to that found *in vivo*. Sedlin [36] showed that the length of time (on the order of minutes) a human bone specimen is allowed to dry in room air is an important factor in bending-strength determinations. Some investigators attempted to solve this problem by controlling the humidity of the test environment, while Sweeney and associates [41] defined "wet" bone as bone stored in 50 per cent alcohol and water and then tested within one hour after removal from the liquid. The one unifying finding throughout these experiments was that as bone dried, the strain to failure decreased. In our laboratory [6] when the specimens were kept wet with a saline drip the amount of strain (especially plastic strain) sustained before fracture in tension was found to be much greater than that previously reported. Since this method of keeping the specimen wet gave higher strains to fracture and the stress-strain curve had a positive but decreased slope after the yield point was reached, we also found higher ultimate tensile strengths than those found in previous tension tests on wet bone (Table VI and Fig. 3). Complete drying of bone has been shown to increase its tensile strength over that of "wet" bone [13,15,18,30,31,36,37]; however, we believe that specimens which have been called "wet" have actually been sufficiently dry to eliminate the plastic deformation portion of the stress-strain curve yet not influence the initial portion of the stress-strain curve.

The elastic modulus reported by our laboratory [6] does not differ from that reported in the literature, but the plastic deformation and ultimate strain are quite different. This is shown in Figure 3, which was constructed using mean values obtained from the authors noted. Since the plastic component of the deformation (that which is responsible for the increased stress and strain to failure) is decreased or eliminated with surface drying, it would seem that maintaining a film of water on the surface of the specimen during the test is necessary to determine the intrinsic strength of bone tissue. Surface-drying times on the order of thirty to sixty seconds are sufficient to reduce the amount of plastic deformation. We [7] postulated that this surface water may compensate for the fact that machining of the specimen creates an external surface from previous internal material. The effect of a water film possibly involves the bonding of the water to high-energy surface molecules. Whatever the mechanism, however, the ability of bone tissue kept wet to display this plastic deformation, and therefore greater ultimate strength and much larger energy absorption before failure (area under stress-strain curves), should be a prominent consideration in future strength testing.

TABLE VII

COMPRESSIVE STRENGTH

Bone Species and Authors	Ultimate Stress ($\times 10^6$ N/m²)	Load Direction	Comments
Human			
Dempster and Liddicoat, 1952	131 106 117	Longitudinal Transverse Radial	Femur, tibia and humerus mixed, dry rewetted; no strain rate given
Yokoo, 1952	159 ± 4.1	Longitudinal	Femur; low strain rate
Bovine			
Sweeney and assoc., 1965	219 153	Longitudinal Transverse	Femur stored in alcohol, no histology reported; no strain rate
Burstein and assoc., 1972	283 ± 24.9	Longitudinal	Femur, plexiform; strain rate: 0.1 sec^{-1}

Fatigue

When a material or structure is subjected to repetitive loads, it quite often fails at a load well below that necessary to produce failure by single application in the same loading configuration. The failure is then called a fatigue failure. This mechanism of failure of bone and bone material is of interest since many *in vivo* loads are cyclic. Fatigue failure in bone has been investigated both on the material and on the structural level. Evans and Lebow [19] machined specimens from the cortex of fresh human tibiae and subjected them to cyclic positive and negative displacements as cantilever beams (a type of loading that subjects the surfaces of the beam to alternate tension and compression). Using equal tensile and compressive loads and subjecting all specimens to the same load level, these workers determined the fatigue life for specimens from different tibiae and from different quadrants within the same bone. They found a very wide range in the number of cycles necessary for failure with a mean value of approximately two million cycles. Specimens from the middle third and posterior quadrant had the greatest fatigue life. There was no consistent relationship found between fatigue life and age.

Tests of this type are quite difficult to apply *in vivo*, since with long fatigue life the repair process may make a significant contribution to the failure properties. Seireg and Kempke [38] considered this and applied repetitive bending loads (which relaxed to zero) on the hind limbs of rabbits. They varied the magnitude of this bending load and found the

TABLE VIII

SHEAR STRENGTH

Bone Species and Authors	Shear Strength ($\times 10^6$ N/m²)	Comments
Human		
Hazama, 1956	53.1	Femur; torsion; low strain rate
Ibuki, 1964	82.4	Femur; single shear; no strain rate reported
Bovine		
Bando, 1961	43.1	Femur, histology not reported; torsion; low strain rate
Ibuki, 1964	89.2	Femur; single shear; histology not reported; no strain rate reported
Sweeney and assoc., 1965	58.6	Femur stored in 50 per cent alcohol, histology not reported; torsion; low strain rate

TABLE IX

BENDING STRENGTH

Bone Species and Authors	Bending Strength ($\times 10^6$ N/m^2)	Comments
Human		
Tsuda, 1957	157	Femur; no strain rate reported
Sedlin, 1965	164 ± 29	Femur; low strain rate
Sedlin and Hirsch, 1966	181	Femur; low strain rate
Bovine		
Kimura, 1952	238	Femur; no strain rate reported

number of cycles necessary to cause fracture under the different conditions. If the maximum bending force was kept below 40 per cent of the force necessary to cause failure in one cycle, they found that the tibiae could be loaded until an arbitrary end point of one million cycles was reached without failure. A distinct change in the load necessary to produce fracture was found at 4,000 cycles. For a fracture to occur with less than this number of cycles a much greater force had to be applied to the limb. The authors postulated that this was the point at which muscle fatigue took place and thereafter the muscles could no longer play a protective role. The frequency of loading was also varied and it was found that the number of cycles to failure was independent of the frequency. A number of specimens were removed from the test apparatus just prior to the predicted fatigue failure and were studied by light microscopy and microradiography for evidence of microcracks or fractures. Since none could be found and the failure of other specimens was accompanied by an audible crack, the authors proposed a brittle mechanism of failure. True fatigue failure is not by a brittle mechanism, however, and only examination of the fracture surface can distinguish between the two.

Fracture Initiation

The point at which a fracture begins or a crack initiates and then propagates is of interest for several reasons. Since bone is a complex composite material, the way in which its constituents are incorporated into the whole at all levels of organization may determine the mode or mechanism of failure. Does the incorporation of hydroxyapatite crystal into the organic phase serve as a nidus for initial crack formation? Is the matrix the "weak link" in the material and therefore the site of fracture initiation? Or, is fracture initiated interlamellarly with a special proclivity for the cementing lines about the secondary osteons? The answers to these questions are bound ultimately to the microscopic structure of the tissue, since with changes in microstructure caused by growth, remodeling, age, mineralization, or pathological condition, changes in the sites of initiation mentioned previously are found.

The orientation or topographical features of a fracture surface can give a great deal of information on the fracturing process. If the stress conditions are known, the stresses on the plane of fracture may be determined by an elastic analysis [29] and the test can then be discussed as a measure of either normal (tension or compression) or tangential (shear) strength. The fracture surface may also give information concerning the energy absorption by the material and the work necessary to drive the crack through the material. Flat smooth surfaces usually result from a quickly traveling crack or a brittle fracture process with low values for energy absorption. Rough surfaces with pull out of constituents usually entail a slower fracture process with more energy absorbed and therefore more needed to drive the crack forward. The natural inclusions and holes in the material necessary for the biological function of the tissue also play a role in determining the initiation and course of the traveling crack.

The nature of the tensile failure when tibial specimens were loaded in a direction parallel to the long axis of the bone was studied by Dempster and Coleman [13]. They found that the plane of fracture was oriented at an angle of approximately 45 degrees to the direction of pull and therefore had a shear component. Microradiographs showed areas which the authors stated were suggestive of fibrous tearing or fraying of a "fibrous matrix." They also noted that there were more secondary cracks and "rarefied slip lines" in their wet specimens than in their dry-tested ones. Their observations are hard to interpret, since they used dried rewetted bones from a museum collection for their specimens and the drying process itself may very well have caused cracks. The courses of the cracks were not influenced by the degree of calcification of the osteons, nor did the cracks seek out the vascular spaces.

When tibial specimens were loaded in a direction perpendicular to the long axis of the bone, a "simple cleavage" with less fibrous tearing and lower ultimate strength was noted. This fracture line tended to follow the curvatures of the cement lines around haversian systems, and only rarely did the fracture line cross an osteon passing through the haversian canal. Based on these findings, Dempster and Coleman postulated that interlamellar cement lines are the weakest structural feature of secondary haversian bone.

Aoji and co-workers [1] also found many cracks along the boundaries between haversian systems and interstitial lamellae, but this distribution of the cracks followed tension tests on specimens of femoral cortical bone loaded parallel to the osteons. These authors also noted that the fracture plane was a "perfect tensile failure," being perpendicular to the direction of the load, and had no shear component. When the loading mode was compression they found that the fracture plane was oriented at an average angle of 60 degrees to the load axis and that there were many interlamellar cracks.

The stress condition causing fractures along planes at 60 degrees of the load axis is a combination of compression and shear. Thus, no simple condition can be stated to be the "cause" of the fracture. Rather, the complex interaction of the transversely isotropic material and the stress field results in failure along the 60-degree planes. Such failures are also seen in other materials.

Compression specimens studied by Dempster and Liddicoat [14], when loaded along the long axis of the bone, failed in a microbuckling manner, the manner often seen in man-made fibrous composite materials. This type of failure occurs when portions of the specimen acting as a structure become unstable and assume other equilibrium positions (resulting in kinking or bowing). These authors suggested that this phenomenon was due to a shear weakness along a plane parallel to the fibers of the bone specimen. They therefore must have assumed that there were internal stresses between components of the tissue which had *different* mechanical properties, since in a uniaxial compression test of a *homogeneous* material there is no shear stress in the direction of the load. Microbuckling may play an important role in the compression failure of bone, but the level at which it occurs (that is, osteonal buckling, lamellar buckling, or collagen-fiber buckling) must be determined.

The failure of bone specimens under a torsional load was attributed to tension weakness by Sweeney and co-workers [41] and by Dempster and Liddicoat [14]. They found that the fracture surface produced by a torsional load was helical in nature, and since this corresponded to the plane of maximum tensile stress they considered the failure to be tensile in nature. However, careful study in our laboratory [6] of this ostensibly oblique surface revealed the presence of an initiating shear step surface parallel to the axis of the torque (Fig. 6). Using both square specimens of cortical plexiform bovine bone and whole dog femora, we found [28] that the initiation of the fracture is in the region of a vascular network in the plane of maximum shear stress which is also parallel to the axis of the torque. Be-

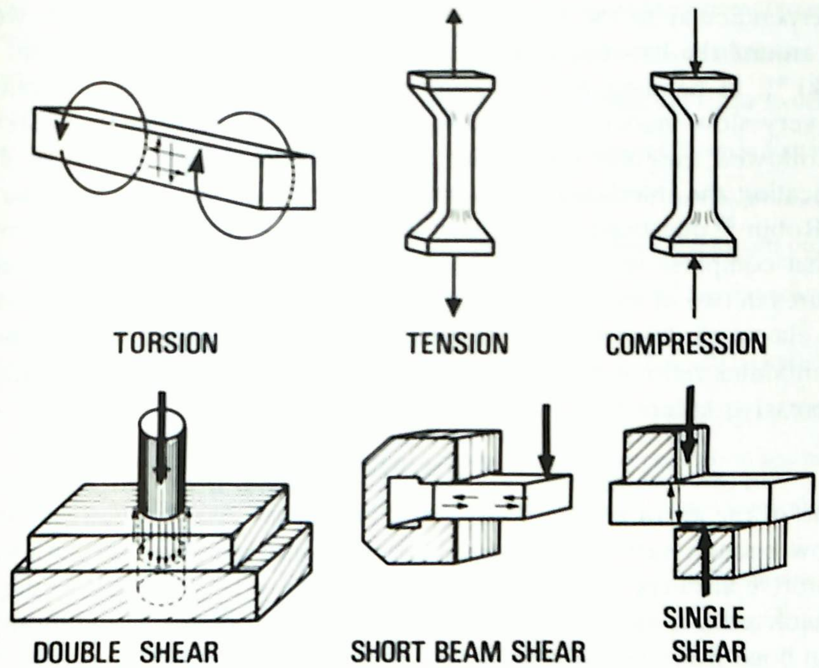


FIG. 5

The three most common material-testing configurations are torsion, tension, and compression. Other methods used to test shear strength besides torsion include double shear, short beam shear, and single shear.

cause of these findings, we maintain that the torsion test is a good measure of the shear strength of the material.

The mode of failure during loading in the bending configuration was studied by Sweeney and co-workers [41]. They found that they could initiate microcracks in bovine bone without producing complete fracture. These fine cracks were first detectable at 60 per cent of the load at failure and were located predominately between lamellae. The crack that produced final fracture was first observed at 70 per cent of the failure load. The increase in load necessary to complete the crack and fracture the specimen represented absorption of energy by the specimen or the toughness of the material. Maj and Toajari [32] bent rectangular prisms of cortical bone and found that when the long axis of the speci-

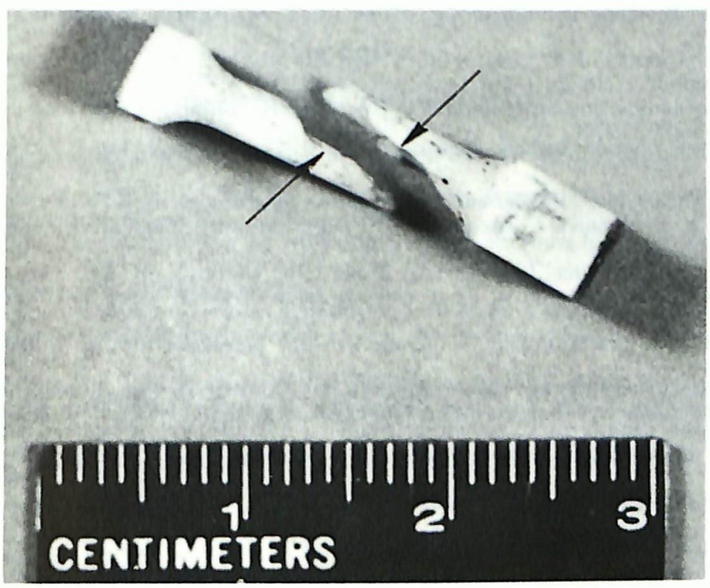


FIG. 6

Torsionally induced fracture in a solid, round bone-tissue specimen. Notice the longitudinal portion of the crack in the middle of the spiral. This longitudinal surface is the initial shear crack, which is then followed by the 45-degree spiral (tension crack).

mens was perpendicular to the long axis of the bones from which they were obtained, cracks went around the haversian systems rather than through their central canals.

Piekarski [34], in bending tests of beams with a triangular cross section, produced cracks with very slow velocities of propagation. The fracture surfaces showed that the cracks had followed irregular paths with whole osteons being pulled or dissected out, further implicating the interlamellar or cement-line weakness of the material. Recently Simkin and Robin [39] questioned the role of tensile or shear failure in bending, raising the possibility that compressive fracture occurs. Their mathematical model predicted compressive failures in two of the four specimens they tested. Their analysis, however, used a compressive elastic modulus which was one-third the tensile modulus. As previously discussed, this modulus ratio does not seem to be likely, or even possible, and their prediction of compressive failure in bending therefore does not seem realistic.

Conclusion

The state of the art of testing bone-tissue properties has reached scientific maturity. There are now enough sufficiently skilled workers interested in the area to produce the required definitive answers. Among the investigations to be expected in the future would be studies characterizing all the necessary elastic, plastic, viscous, and failure characteristics of human bone from the same specimen population. In such studies we would expect bone to be treated as a transversely isotropic material. Another required study would look at donor age as a variable, while still another study, also needed, would characterize the changes in the properties of bone tissue occurring with various metabolic diseases.

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Appendix

Supplemental Bibliography

1. ABENDSCHEIN, WALTER, and HYATT, G. W.: Ultrasonics and Selected Physical Properties of Bone. Clin. Orthop., **69:** 294-301, 1970.
   Ultrasonic measurements were good but for the statically determined properties a bending test with a strain gauge glued to tension surface was used.
2. AMTMANN, EDUARD: The Distribution of Breaking Strength in Human Femur Shaft. J. Biomech., **1:** 271-277, 1968.
   Embalmed, dry-bone specimens were used.
3. BANDO, T.: Studies on the Torsion Test of Compact Bone. J. Kyoto Pref. Med. Univ., **69:** 633-648, 1961.
   Unknown histology, no strain rate reported, and no correction in ultimate stress for non-linearity of stress-strain curve.
4. BONFIELD, W., and LI, C. H.: Deformation and Fracture of Bone. J. Appl. Phys., **36:** 869-875, 1966.
   Bovine bone, the histological features of which were unknown, was used. Bone seems to have been essentially dry (in view of low ultimate strains) and the wide temperature range (−200 to +900 degrees centigrade) used during testing suggests possibility of a change in the material to something other than native bone.
5. DEMPSTER, W. T., and LIDDICOAT, R. T.: Compact Bone as a Non-Isotropic Material. Am. J. Anat., **91:** 331-362, 1952.
   Edge effects on small cubes tested in compression could have caused large errors in Young's modulus and stress at failure. Movement of testing machine platens was used to measure strain.
6. DEMPSTER, W. T., and COLEMAN, R. F.: Tensile Strength of Bone Along and Across the Grain. J. Appl. Physiol., **16:** 355-360, 1960
   Dry rewetted bone was used. These bending tests for tensile strength were based on the assumption that there was linear elasticity. They were also plate tests rather than beam tests, but the appropriate analysis for a plate test was not used.

7. EVANS, F. G., and LEBOW, MILTON: Regional Differences in Some of the Physical Properties of the Human Femur. J. Appl. Physiol., **3:** 563-572, 1951.
Embalmed material was used.
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9. EVANS, F. G., and VINCENTELLI, RAÚL: Relation of Collagen Fiber Orientation to Some Mechanical Properties of Human Cortical Bone. J. Biomech., **2:** 63-71, 1969.
Embalmed bone was used and the collagen-fiber orientation was determined by polariscopic studies.
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Bovine bone fixed in formol and ethyl alcohol was used for small compression specimens. Three directions were studied but edge effects were probably large.
11. HIRSCH, CARL, and DASILVA, ODILIO: The Effects of Orientation on Some Mechanical Properties of Femoral Cortical Specimens. Acta Orthop. Scandinavica, **38:** 45-56, 1967.
Bending tests were used to study the variation in Young's modulus with respect to orientation. Comparison of zero degrees (longitudinal) and 90 degrees (transverse) may be valid in a relative sense but the in-between values were less meaningful since anisotropic theory was not used.
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13. MCELHANEY, JAMES; FOGLE, JOHN; BYARS, EDWARD; and WEAVER, GEORGE: Effect of Embalming on the Mechanical Properties of Beef Bone. J. Appl. Physiol., **19:** 1234-1236, 1964.
14. MAJ, GIORGIO, and TOAJARI, EZIO: Osservazioni sperimentali sul meccanismo di resistenza del tessuto osseo lamellare compatto alle azioni meccaniche. Chir. org. mov., **22:** 541-557, 1937.
Small parallelipipeds, tested in bending, were more like plates than beams. Plate analyses of data were not done.
15. MATHER, B. S.: Correlations Between Strength and Other Properties of Long Bones. J. Trauma, **7:** 633-638, 1967.
Whole bones were used in bending tests.
16. MATHER, B. S.: The Symmetry of the Mechanical Properties of the Femur. J. Surg. Res., **7:** 222-225, 1967.
Whole bones were used in bending tests.
17. MATHER, B. S.: The Effect of Variation in Specific Gravity and Ash Content on the Mechanical Properties of Human Compact Bone. J. Biomech., **1:** 207-210, 1968.
Whole bones were used in bending tests.
18. SMITH, J. W., and WALMSLEY, R.: Factors Affecting the Elasticity of Bone. J. Anat., **93:** 503-523, 1959.
Samples of human bone were tested in bending with the loads allowed to remain for two minutes before deflection readings were taken.
19. VOSE, G. P.: The Relation of Microscopic Mineralization to Intrinsic Bone Strength. Anat. Rec., **144:** 31-36, 1962.
Dried bone specimens from tibiae were used.
20. VOSE, G. P., and KUBALA, A. L., JR.: Bone Strength — Its Relationship to X-Ray-Determined Ash Content. Hum. Biol., **31:** 261-270, 1959.
Bending tests on dried whole human femora were used.
21. WEIR, J.B.DEV.; BELL, G. H.; and CHAMBERS, J. W.: The Strength and Elasticity of Bone in Rats on a Rachitogenic Diet. J. Bone and Joint Surg., **31-B:** 444-451, Aug. 1949.
Dried whole bones were studied.