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MORPHOLOGY AND FRACTURE OF BONE

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ABSTRACT

An attempt has been made to correlate the brittle and ductile behaviour of bone to its microstructure. Neglecting the contribution of liquids, cortical bone may be considered as a composite material consisting of a mineral and an organic phase. A more sophisticated model should incorporate liquids as structural members and also consider an organic phase which in certain circumstances, may behave like a liquid. The latter model is more realistic and more applicable to the interpretation of crack propagation in bone. In this model, the lamellar structure of osteons includes concentric cylinders comprising a high volume fraction of a polycrystalline mineral separated by an organic-mineral aggregate. This interlamellar aggregate is the major contributer to the ductility of bone. With advancing age, this aggregate becomes radially bridged by calcified polycrystalline tissue rendering the bone brittle. Not all osteons, however, calcify: some of them become remodelled and form young osteons, with the same characteristics as in young bone. Thus the brittle or ductile behaviour of bone depends not so much on the age of an individual as on the ratio of old to young osteons in the bone.

INTRODUCTION

Bones are structural components of the body and transmit predominantly compressive, torsional and bending stresses. By offering high resistance to deformation, they also protect more vulnerable organs from injury. Both mechanical functions should be performed with the minimum weight of the bony components and the maximum efficiency. It is realized only today in engineering that one of the best energy absorbing mechanisms is a liquid filled porous material [1]. Nature has designed the cranium to protect the brain from external injury in precisely this manner. Liquids in bone have, of course, another function recognized a long time ago; they serve as a transport medium for the nutrients which have to be supplied to cells, and for the waste products which have to be carried away. Soft tissue such as muscles, tendons and ligaments transmit tensile stresses and apply compressive loads to bone.

Trabecular bone, which has a cellular, porous structure and is often referred to as cancellous or spongy bone, is designed primarily to absorb energy and transmit compressive stresses. The compressive strength of the trabecular bone is lower than the strength of the cortical bone which has a relatively solid structure, hence this bone has to be thicker. The ends of long bones (epiphysis) which support the bearing surfaces of joints have to absorb energy, but also have to transmit compressive stress

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to the rest of the bone. Thus it is quite logical that they are thicker and are made from the spongy trabecular bone rather than from cortical bone.

The fact that long bones are thinner in the middle (diaphysis) but still maintain the same strength has another reason. The distribution of weight bearing material is such that the compressive stress remain the same throughout the whole length of the femur (Figure 1). However, the smaller size of bone in the middle reduces the strain in bending, thus increasing its ability to bend and reducing the probability of brittle fracture. With age, the proportion of solid bone decreases: the walls of long bones become thinner and cannot resist the same stresses. Nature compensates for this deficiency by increasing the outer diameter of bones, thus making them structurally stiffer. However, this also increases the amount of strain in the outer fibres on bending and since bone is a strain-sensitive material, it becomes markedly more vulnerable to brittle fractures. It is also interesting to examine the cross-sectional shape of the femur. In engineering applications, the shape of the shaft depends on the type of stresses which the shaft is designed to resist. Thus, a cylindrical shape is the best for resisting torsional stresses and a square shaft is optimal for resisting bending stress applied parallel to its sides. The shape of the shaft of the femur is triangular with well rounded corners and the apex of the triangle points to the back of the body. This is the compromised shape for resistance to both torsional and bending stresses.

Microstructure of cortical bone

Figure 2 illustrates schematically the microcomponents of a cortical bone. It consists of four constituents:

1. the periosteal bone having lamellar structure following the outer curvature of the cortex,
2. the endosteal bone, also lamellar but the radius of the lamellae corresponding to the inner radius of the cortex,
3. Haversian bone,
4. interstitial bone.

The whole length of a shaft of a long bone is surrounded by the periosteum on the outside and endosteum on the side of the marrow cavity. Periosteum and endosteum is made up of connective tissue and is packed with cells: it has a very important biological rather than mechanical significance. The rest of the compact bone is also permeated with cells located in the lacunae or cavities, which have an approximately elipsoidal shape. The shape and distribution of the lacunae, a discontinuity in the compact bone, has a mechanical significance when the resistance of bone to fracture is considered. Cells located in lacunae receive their supply of nutrients from the Haversian and Volkman's canals through the network of canaliculi.

Figure 3 is a transverse cross-section of bone polished with metallurgical techniques and examined under a reflected light microscope. This photomicrograph clearly shows that the shaft of a cortical bone does not have a homogeneous structure. The outer rim (periosteal bone) clearly shows its lamellar structure and it does not contain blood vessels in Haversian systems like the centre of the cortical bone and thus its properties are also different. The endosteal bone, on the inner rim has a very similar structure to the periosteal bone and therefore also similar properties.

The Haversian bone, between the rims, consists of a network of blood and lymphatic vessels crisscrossing bone in hollow canals (Figure 4). Each canal is surrounded with concentric lamellae forming an "osteon" or a "Haversian system". Figure 3 shows the ends of osteons which are standing up in relief with respect to the rest of the bone which can be considered a matrix and is called an "interstitial bone". The resistance to abrasion of osteons seems to be better than the interstitial bone indicating that they are stronger. This indeed is the fact, and it has been shown [3] that a propagating crack avoids osteons if they are oriented parallel to the applied tensile stresses. Figure 4 is a photomicrograph taken with the scanning electron microscope showing a longitudinally cleaved section of a long bone. It should be noted that Haversian and Volkman's canals are not as perfectly symmetrically oriented as illustrated in Figure 2. However, the orientation of Haversian canals is predominantly parallel to the axes of long bones; it has also been shown that Haversian canals follow a spiral path along the cortex. The most profound effect on the mode of fracture of long bone is its lamellar character and therefore, the microstructure of the lamellae need to be thoroughly investigated.

The photomicrograph of a thin section of bone recorded in Figure 5 shows more clearly the structural morphology of osteons. It illustrates the concentric position of lamellae around the central canal and a network of interconnecting canaliculi. Figure 6 is the same specimen viewed with polarized light. In this case it is possible to observe in more details the lamellar structure of Haversian systems. The structure of osteons has been compared by Ascenzi and Bonucci [4] to the cross-ply laminates indicating that the fibrous structure of each lamellae is oriented at a different angle to the long axis of an osteon. However, black and white coloring of lamellae may also indicate differences in composition or microstructural arrangement of organic and mineral phases within lamellae.

In order to identify closer structural differences between lamellae, it is useful to examine an osteon with one of the phases removed. Figure 7 shows an osteon with a decalcified surface produced by etching with HCl. The lamellar character is clearly demonstrated indicating that ridges standing up in relief must have been rich in organic phase and valleys between them consisted predominantly of a mineral phase. Figure 8 illustrates an osteon from the fractured surface of bone treated with ethylene diamine. This treatment removed completely an organic phase illustrating that alternate lamellae are made of a continuous hydroxyapatite phase, possibly in polycrystalline form. The concentric cylinders of hydroxyapatite are separated from each other by empty spaces with some mineral crosslinking. It may be quite safely speculated from this photomicrograph that the empty spaces were filled predominantly with an organic phase or that the mineral present existed there as a discontinuous phase.

Clearly the above description of the lamellar microstructure of bone must have a great influence on the character, mechanisms, and mechanics of crack propagation through bone.

Material phases in bone

A compact bone is not a homogeneous material. Apart from many discontinuities, the "solid" portion consists of substances which have completely different properties.

The main components which have an effect on mechanical properties of bone may be grouped in the following manner:

1. Crystalline mineral phase - hydroxyapatite
2. Amorphous mineral phase - composite approximating hydroxyapatite
3. Crystalline organic phase - collagen
4. Amorphous organic phase - protein molecules in the form of gels and sols
5. Liquids

The effect of the above phases on the overall properties of bone depends on the amount of phases present, their properties and their distribution. At slow rates of deformation the effect of liquids and sols on the load bearing capacity of bone is very small and it may be neglected, if it can be assumed that they can flow freely in and out of a porous portion of bone. This assumption would justify the replacement of liquids and sols in the list of structural components with pores.

Further simplification can be made by grouping together crystalline and amorphous mineral phases for the studies of an elastic modulus of the composite. The elastic modulus is a structure insensitive property and it can be safely assumed that it has the same value in the crystalline as in the amorphous mineral phase. The amount of the amorphous organic phase is very small and for the present purpose it may be categorized along with collagen.

This would reduce the list of components in bone to three:

1. Mineral phase
2. Organic phase
3. Pores (filled with liquids)

This simplification allows bone to be visualized as a composite resembling more familiar engineering materials. The properties of such a composite (bone) depend on the amounts of the components present as well as on the properties of the individual components.

Mineral phase

It has been demonstrated that the mineral phase in bone is that of apatite, more specifically that of hydroxyapatite $CA_{10}(PO_4)_6(OH)$. It has hexagonal crystal structure with unit cell dimensions a = b = 9.47 x 10^{-4}µm, c = 6.88 x 10^{-4}µm. Some investigators [5,6] suggest that the crystals are rods having diameters of some 20 to 70 x 10^{-4}µm and extending in length from 5 to 100 x 10^{-3}µm.

The fact that bone remains in one piece and even displays a very good strength after an organic phase is removed [7,6] indicates that the mineral phase must be continuous. It may be postulated that mineral crystallites are joined end to end forming a network of long and very thin, heavily crosslinked fibres.

The nucleated crystallites are oriented with their "c" axes parallel to the long axis of the collagen fibre and the preferential direction of growth of these crystals is also in the same direction producing the rod-like appearance of the inorganic phase. When growth occurs it is very difficult to distinguish between individual crystals which are joined

together and thus obscure the initial pattern.

The mechanical properties of hydroxyapatite are very little known and it will be necessary to make a series of assumptions and speculations.

Bhimasenacher [9] reported that the elastic modulus of apatite was E = 109,627 MPa; Currey [10] estimated it at E = 172,369 MPa. Sweeney [8] and Mack [7] have tested properties of bone with the organic phase removed. Typical values of these experiments are shown in Table 1.

The values in Table 1 can be converted into values for a solid mineral assuming that porosity is about 60% and relationships developed for porous ceramics apply to the porous hydroxyapatite. According to Ryskewitch [11] strength (σ) decreases exponentially with porosity

$$\sigma = \sigma_o \exp(-nV) \tag{1}$$

where n is in the range 4 - 7 and V is the volume fractions of porosity. The elastic modulus, according to Coble and Kingery [12] changes with porosity following the expression

$$E = E_o (1 + 1.9\ V + 0.9\ V^2) \tag{2}$$

Pores

The porosity of a cortical bone consists of blood vessels (Haversian canals) interconnected by canaliculi with lacunae. According to Frost [13] this occupies 13.3% of volume for a human bone. However, true porosity must be much higher because it is possible to evaporate liquids from bone yielding value for porosity equal to 18%. In addition, it should be realized that liquids which normally fill pores could not all be evaporated. The solid portion in the liquids consists of organic and mineral ions plus large amounts of various protein molecules. Therefore, the actual value of porosity will be well above 18% but this figure has not been estimated and it is not used for this study.

Applying the above relationships, the mechanical properties of hydroxyapatite in tension can be estimated as follows:

σ = 107 MPa E = 40670 MPa

However, the absolute value of the modulus of bone apatite still remains a matter for conjecture and, for the present purpose, it may be assumed that E = 69000 MPa.

Organic phase

This phase consists approximately of 99% of collagen. Collagen is a product of the polymerization of amino acids - the main groups are hydroxylsine and hydroxyproline (Figure 9). The single polymerized polypeptide molecule has a left-handed helical shape and this chain in turn twists into a right-handed super helix. Three super helixes twisted into one form the fundamental unit of the collagen structure; this has been named procollagen [14]. The collagen fibril consists of a longitudinal arrangement of procollagen units arranged in such a way that the fibril shows a periodicity of 64-70 x 10^{-3}µm in its structure - this is caused by the uneven distribution of electronic charges along its length. The periodicity of the structure has

been demonstrated by precipitation of collagen fibrils from acetic acid and by X-ray diffraction analysis [15]. The regularity and symmetry of the collagen structure fully justifies the application of crystalline terms to the organic component of bone structure. The distribution of fibrils in the matrix is not, however, perfectly symmetrical. The thicker fibres, being a "back bone" of the structure follow the same general direction and are responsible for the anisotropic character of the material, the fine fibres form a network which interconnects the main branches and displays less directionality. The length of the fibres has no particular limit and the network is continuous so that, even after removal of the inorganic phase, bone still maintains its original shape. Little [16] has also demonstrated the existence of tubular fibres of collagen by stereoscopic electron micrography of individual fibres. The variation of thickness of collagen fibres is attributed to the cementing of fibrils by the amorphous phase of mucopolysaccharides.

Deformation of collagen

The elastic properties of a crystalline solid arise from the short range interatomic forces and it may easily be shown that the elastic modulus is obtained from an expression of the form

$$P = \frac{d\phi\ (u)}{du} = \frac{d^2\phi}{du^2}\ u \qquad (3)$$

where ϕ (u) is the bond energy at the displacement u and P is the interatomic force. This indicates that the controlling factor is a change of energy. Within the lattice under zero load each atom is in position of minimum energy, on application of a load the atoms are displaced to positions of higher energy and, on removal of the load, they return to the configuration of minimum energy.

Early molecular theories of rubber explained its elasticity by considering the behaviour of the spring-like shape of its molecule. The type of coiling of a long molecule depends on its morphology. The composition of natural rubber, for example, and gutta-percha are exactly the same but their molecular structures differ only in the position of one hydrogen atom. This small change in structure makes all the difference in their properties. Natural rubber is soft and capable of large elongations while gutta-percha is hard and brittle.

The molecule of a natural rubber has too many atoms on one side and hence has a tendency to curl forming coils. The coiled structure explains qualitatively the stress-strain curve of elastomers but does not provide a complete theory. There are two main contradictions. Firstly, the elastic modulus of an elastomer increases with increasing temperature; secondly, its coefficient of thermal expansion is positive up to a few percent of elongation and then it becomes negative. These apparent discrepancies were justified by many workers by taking into consideration the change in entropy, i.e., the change in the amount of order and disorder in the molecular arrangement. The deformation of rubber differs from that of other materials by the fact that its symmetry, or the amount of order in the molecular structure is increased when the elastomer is put in tension. In other materials the disorder (i.e., entropy) increases both in tension and compression.

Based on the observation that fibrils of collagen have a molecular arrangement which is similar to that of rubbers and that the behaviour of muscle was also thermodynamically similar to rubber, [17], Hall [18] performed a series of experiments on an animal tendon (Figure 10). Hall's data from tests carried out at 35°C and pH 7 did not verify the rubbery behaviour of collagen in tendon. However, the same experiment repeated at pH 1.75, where collagen fibres contracted over 25%, showed that an initial extension up to the normal length of fibre at pH 7, was accompanied by a decrease in entropy; i.e. rubbery behaviour (Figure 11). The effect of swelling in the acid solution was obviously a retraction of the molecular chains into a less orderly arrangement. On subsequent stretching, the molecular arrangement becomes more orderly, entropy decreases until the original length at pH 7 is reached, and further stretching follows the pattern as in the first experiment where the energy effect was predominant.

Similar swelling of a protein molecule in acid was reported by Jordan, Lloyd and Garrott [19] in rat-tail tendon. Acid causes the collagen molecule to combine with hydrogen ions to break some of the saltlinks; thus the breakage of cross-links allows collagen chains to retract fully resulting in an increase of entropy. Subsequent stretching of the acid-swollen fibre results in molecular realignment and decrease in entropy. The arrangement of collagen in bone is not known because of its intimate contact with hydroxyapatite. It is very unlikely that collagen molecules in calcified bone are more coiled than in tendons and therefore its behaviour should also not have a "rubbery" character. A detailed thermodynamic study of bone was carried out by Piekarski [20] and energy and entropy contributions to the deformation of bone were similar to Hall's results for animal tendon.

Static mechanical properties of collagen

Direct measurement of the properties of collagen is difficult because the fibres are generally too small to handle. However, tendon is composed mostly of collagen and the properties of tendon have been measured by various researchers with the results shown in Table 2. Typical stress-strain curve for an animal tendon, which is predominantly made of collagen is illustrated in Figure 12.

The tensile stress-strain curve is nonlinear and the elastic modulus reported above was calculated from the most nearly linear part of the curve - this lay generally between strains of 10 and 20%. Changing the rate of strain does not produce any apparent change in properties but the scatter in the results is so large that it may mask any effects due to this variable. The effect of testing temperature is not known precisely although Hall [21] has reported that the creep rate of tendon changes considerably if the temperature is changed from 20 to 35°C.

Load relaxation and recovery experiments further confirm the viscoelastic character of tendons and its behaviour is similar to that of a cross-linked thermoplastic polymer. Figure 13 illustrates a typical experimental curve for tendons.

Tendon also displays some mechanism of plastic deformation. This may occur by sliding of chain molecules with respect to each other, as in rubbers, or by plastic flow in the amorphous parts of the organic phase (amino acids and mucopolysaccharides) which interpenetrate and surround the collagen fibres. The precise mechanism is still not known.

Rheologically, the behaviour of collagen may be illustrated by the model shown in Figure 14. It indicates that in the normal range of loading, collagen displays fully recoverable viscoelastic deformation. However, when overloaded it will produce nonrecoverable plastic flow. This behaviour has a profound clinical significance indicating that by application of therapeutic treatments, tendons or the collageneous part of hard tissue may be permanently stretched.

Properties of bone as a composite material

Perhaps the simplest and most useful model to predict the elastic modulus of bone from the properties of components is the Hirsch model [23]

$$\frac{1}{E} = 0.9\left(\frac{1}{V_m E_m + V_o E_o}\right) + 0.1\left(\frac{V_m}{E_m} + \frac{V_o}{E_o}\right) \qquad (4)$$

Assuming further that the volume fractions of mineral and organic phases are equal, it is possible to calculate the elastic modulus of bone.

The value obtained is:

$$E = 16754 \text{ MPa}$$

This value is approximately correct, assuming that the test was carried out at low strain rates.

The other approach to the modelling of bone as a composite material is to assume that it consists of two different phases: solid and liquid. A very low elastic modulus of the organic phase indicates that it cannot contribute very much to the load bearing capacity of bone, unless its displacement is restricted by the mineral phase. Such a behaviour corresponds more to viscous deformation of sols, gels and liquids than rigid solids. In such a case, the organic phase will transmit hydrostatic forces at high strain rates, and at low strain rates, at which viscous flow would have sufficient time to relieve hydrostatic pressure, its contribution to the load bearing capacity would be negligible.

Assuming that this postulate is true, the value of the elastic modulus in tension of porous hydroxyapatite, as reported by Mack (Table 1), is of the right order of magnitude. It is about the same as for bone tested at low strain rates. The increase of the elastic modulus with strain rates has been reported by many investigators [24,25] and it can easily be the result of the increase of hydrostatic pressure in liquids and in the organic phase.

Inclusion of liquids as an important microcomponent of bone requires an introduction of a new model for such a complex composite material as a cortical bone. Hirsh's model assumes a uniform distribution of mineral and organic phase. Coefficient (0.9) in equation (4) indicates that 90% of phases are arranged parallel to the long axes of bones and, therefore, a simple mixture rule can be applied with close approximation in order to determine properties of the composite. However, Figure 8 indicates that the arrangement of phases in the lamellae of an osteon is not as simple as to be treated by a mixture rule. Studies of Ascenzi and Bonucci [4,6] have already revealed that the fibrous arrangement of phases is not always parallel to the long axis of an osteon but it even differs from one lamellae to another. Treatment of an osteon with ethelyne diamine has

removed most of the material from between concentric cyclinders constituting lamellae of osteons. This indicates that the material removed was either all organic or an aggregate of a continuous organic phase with the discontinuous crystallites of a mineral phase.

Based on this observation it is possible to model an osteon as a series of concentric cylinders separated from each other with the liquid phase. An approximation of an organic material or organic and mineral aggregate to liquid is reasonably justified if it is remembered that structure, mechanical properties and mechanical behaviour of collagen most closely resembles behaviour of rubbery materials for which the Poissons ratio is generally assumed to be equal to 0.5, i.e. upon compression no change of volume occurs and, therefore forces transmitted through such a medium are comparable to the transmission of the hydrostatic pressure through liquids.

A model of an osteon is illustrated in Figure 15. From such a model it is possible to predict the hydrostatic pressure in each liquid annulus [26]. Since Haversian canal is interconnected with lacunae (Figure 2 and Figure 5) by the multitude of canaliculi distributed radially through all lamellae, it may be assumed that after initial increase in pressure a flow would occur to reduce it to the normal blood pressure in the central lumen of an osteon. Thus the total volume change for a liquid annulus is

$$\Delta V_{total} = \Delta V_{radial} + \Delta V_{lateral} + \Delta V_{flow\ in} - \Delta V_{flow\ out} \qquad (5)$$

Equation 5 can be written for each liquid annulus and the resultant set of simultaneous linear algebraic equations can be solved for the unknown values of the hydrostatic pressure. Obviously a series of assumptions had to be made in order to determine such pressure distribution. The assumptions of dimensions, properties of solids and liquids, the mode of flow and the resultant pressure distribution through the lamellae of an osteon are illustrated in Figure 15.

Stress concentrations in bone

The most comprehensive study of stress concentrations in bone has been made by Currey [27]. There are macroscopic and possible microscopic stress concentrators in the long bones. The centre of long bones (diaphysis) is smooth, but the epiphysis or ends of long bones have all kinds of lumps, steps, grooves and notches which originate from attachments of tendons or cartilage forming articulating surfaces. Fortunately the highest bending stresses, i.e., the highest tensile stresses act on the outer surface of diaphysis of bones which are free of irregularities. It should also be pointed out that the epiphysis is made up of a cancellous (trabecular) bone which is much less prone to propagate a crack than a solid cortical bone. The same reasoning may also be applied to the outer and the inner surface of the cortex. The inner cortex has a much rougher surface but since it is closer to the neutral axes, it is subjected to lower stresses in bending. Cortical bone contains many discontinuities in its matrix such as Haversian canals, canaliculi and lacunae. Currey considered them as stress concentrations and calculated stress concentration factor for these discontinuities ranging from 2.5 to 7.0 There are, however, several reasons why these discontinuities cannot act as stress concentrators but would rather contribute towards arresting propagating cracks. One reason is that, considering bone as a brittle material, the sizes of discontinuities are much smaller than the critical Griffith crack size [3]. The other

more important reason is the fact that Haversian canals are protected by concentric lamellar bone with weak interfaces which deflect propagation of the crack [3,4,7,28,29,32]. Less obvious interfaces exist around canaliculi and lacunae, Figures 16 and 17. The solid bone always contains a certain amount of porosity as can be observed in Figure 16, however, the canaliculus is surrounded with highly calcified nonporous structure which presumably is much stronger than the rest of the porous bone and also offers a clearly defined interface along which a crack may be deflected. Figure 17 shows a similar condition of a stronger protective envelope around the lacune. Assuming, however, that an external crack would penetrate one of the above discontinuities, the tip of the crack will be sufficiently blunted to reduce the stress intensity factor rather than increasing it.

Finally, the last factor which would make the discontinuities the unlikely stress concentators is their spacial orientation. Most Haversian canals run parallel or at a small angle to the long axis of bone and lacunae which have approximately elipsoid shapes, have their short axis oriented in the radial direction thus reducing further their possible stress concentration effects.

Toughness of bone and the mechanism of crack propagation

Bone is a highly anisotropic material, on both microstructural and submicrostructural levels, consequently it is only logical that the energy to propagate a crack and the mechanism of crack propagation depends on the spacial orientation of a test specimen. It has been shown already in 1966 by Bonfield and Li [30] that the energy absorbed during fracture is much higher in the longitudinal than in the transverse direction. Testing cortical bone at various temperatures (from -200^{0}C to +500^{0}C) they observed a peak of the energy absorption at 0^{0}C for both types of specimens. Although testing bone at this range of temperatures does not have very much biological significance it is of interest from the point of view of the material behaviour and similar investigations were carried out by Armstrong et al [31].

Energy absorption during fracture of bone has been measured by a number of investigators and some of the results are listed in Table 3.

Although the magnitude of the reported results differs, it is evident that energy to fracture decreases with age. It was also observed in this laboratory that even within the same age group the scatter of the results is very large. It may be speculated at this stage that the quality of bone depends to a great extent on the activity of the individual and since this data was not available, the correlation will remain just a hypothesis.

There is also a large number of researchers who have attempted to observe the mechanism of deformation in cortical bone. Ascenzi and Bonucci [37] have tested single osteons in tension and on the basis of stress-strain curves and microscopic examinations concluded that characteristics of deformation depend on the orientation of fibers in individual lamelli. After the elastic limit of fibers has been reached, the interfibrillar cementing substance fails and plastic deformation occurs resulting in a series of microfractures. Although their photomicrographs [37] show distinctly different lamellae (black and white), the lamellar pull-out has not been observed. Most investigators have confirmed the previous findings in this study [3] by identifying two mechanisms of fracture: ductile by the pull-outs of osteons and fibers, and brittle occurring mostly at higher strain rates, by the crack running indiscriminately across microconstituents in a quasi-cleavage manner.

Results almost identical with this study have been reported by Saha in shear [34] and in tensile tests [38]. Precracking specimens before bending tests proved to be a very useful technique for Pope and Outwater [39], in order to obtain a controlled propagation of the crack. They have also attributed "plastic" deformation in bone to the inter-lamellar debonding mechanism. Currey and Butler [35] have also observed that energy absorption of the travelling crack is related to the roughness of the product surface. Comparisons of tensile tests with dynamic tests at strain rates of 133 sec^{-1} were made by Saha and Hayes [40]. A 34% increase in strength was observed in the dynamic tests and high expenditure of energy on the initiation of a crack. Fractography showed also perfectly brittle failures in the dynamic tests.

Fatigue failures of bones

The first experiments showing fatigue fracture of bones were performed in 1922 by Muller [41] who removed a segment from a radial diaphysis of a dog's paw and, after a certain period of time, forcing the dog to walk on it, observed fatigue failure. The numerical value of the fatigue strength of compact bone was first determined by Evans and Lebow [42] in 1957. Since that time a surprisingly small amount of research was allotted to this problem. It was generally recognized that bone fractures under cyclic loading, as it was observed in "walking fractures" in soldiers during the First World War, but it was only in 1960 when Frost [43] made a very remarkable study. He postulated that the human rib cage is loaded cyclically by breathing and ribs should fail in fatigue unless the repair mechanism of the fatigue cracks occurs faster than their formation. Indeed the fatigue cracks have been observed but the repair mechanism still remained a conjecture. Seireg and Kempke [44] in 1969 succeeded under laboratory conditions in producing fatigue failures in tibias of white rats and accummulated data in the form of load-cycle curve. Chamay [45] performed experiments in vitro on dogs' ulna and concluded that in order for the fatigue fracture to occur, bone should be loaded in the "fatigue zone" which he arbitrarily assigned as near the top portion of a stress-strain curve. The first correlation between the microstructure of bone and the fatigue strength was made by Evans and Riolo [46] who reported a positive correlation at the 0.01 significance level between the fatigue life of bone and the percentage of osteons in the break area.

Numerical values of the fatigue life of bones tested by the above investigations is not comparable and is not expected to be comparable, because there are no standards according to which such a study may be conducted. In addition, it should be emphasized that even if there were such standards, values obtained would have no practical meaning unless correlation is made with the fatigue life of a human bone in vivo.

The effect of age on fracture of bones

It has been well established that the metabolic efficiency of organs and organisms declines with age. It is also a well-known fact that old people fracture bones much more easily than children. One of the reasons was mentioned in the introduction to this paper, the other clearly must be in the differences in the microstructure of cortical bone. Posner et al [47] observed a significant improvement in the crystallinity of hydoxyapatite with age. A more organized mineral phase would improve strength of bone but decrease the ductility. Villanueva et al [48] and Frost [49] have related the amount of remodelling in bone to the process of aging. Epker and Frost [50] concluded on the basis of extensive studies, that

there are only minor changes in the kinetics of the formation of the average individual new osteon with aging. Thus it may be postulated that it is not the quality of the individual osteons which decide on the properties of bone but the properties must be related in some way to the average age of osteons in the bone which in turn depends on the age of the person. Currey [27] has speculated that bone cells must have some relationship with the metabolic function of bones, although during the remodelling process osteoblasts die and new bone and new osteoblasts are created by the proliferation of mesanchymal cells; the amount of the original osteoblasts in the bone has an influence on the rate of remodelling. Hattner et al [51] have studied specifically cell behaviour in osteons as a function of age. Perhaps the most significant conclusion reached as a result of their studies was that the amount of protein in the finished osteons "tends strongly to remain constant" independent of age. Chatterji and Jeffery [52] attributed decrease of "pliability" of bone with age chiefly to the physical properties of the apatite crystals. The observations made on cleaved fractures indicated an increase in size of apatite crystals with age.

From the observations made by so many researchers, the following synthesis could be made:

a) The rate of remodelling of bone decreases with age

b) The composition, morphology and size of newly formed osteons are almost the same in young and old bones

c) Aging osteons change their composition somewhat showing the increased amount of mineral phase which exists in the form of larger size crystallites.

New experimental work

It may be concluded from the reports of previous investigators that toughness of bone is related to the mechanism of crack propagation through the microstructural constituents of a cortical bone. The amount of energy absorbed is related to the osteon pull-out mechanism or production of a rough fracture surface. This in turn depends to a large extent on the rate of crack propagation. A ductile-brittle transition has been reported by Hasson and Armstrong [53] indicating that there could be two modes of deformation present in the same specimen and also showing that brittle fracture occurs more frequently at lower temperatures.

In the present experimental study, human bone from subjects of various ages obtained at post mortem or after amputations and preserved in deep freeze has been fractured at low strain rates. In order to observe any possible ductility even in very brittle bones, the Tattersall and Tappin [54] method of three point bending was applied. As a result large differences in the energy absorption were noted (see Table 3) and the characteristics of the load deformation curves indicated the presence of either ductile or brittle material under test (see Figure 18 b,c). The obtained results were so strikingly different that electron scanning micrography was used to determine the mechanism of crack propagation in brittle and in ductile specimens.

Fractography and measurements of energy required to propagate a crack confirmed earlier findings of other investigators [35]. Young bones absorbed more energy and fracture was generally ductile. There were large portions of the fractured surface with pulled-out osteons clearly demonstrating the

mechanism of energy absorption (Figure 19). Specimens from a human bone in the age range between 30-60 years did not produce any consistent results. Some fractures were ductile, some brittle or a portion of the fractured surface would have pulled out osteons and the part immediately adjacent to it would show an almost cleaved surface. Upon more accurate observation it was possible to distinguish osteons with entirely different morphology. Osteons contributing to the pull-out mechanisms demonstrated clear concentric lamellar structure (Figure 20) with large gaps between concentric cylinders. The gaps were as large as in Figure 14 where the organic phase has been removed. The possible explanation may be that the mineral and organic aggregate shrank upon drying in preparation for scanning electron microscope examination. Osteons found in brittle bones did not show such distinctive concentric lamellae (Figure 21). There is some indication that such lamellae existed when an osteon was originally formed but in time radial calcification occurred bridging them into a solid mass. This may be the result reported first by Frost [55] and discussed by Currey [27] that in old bone when osteoblasts die in the lacunae, remodelling from the blood vessel does not always occur, but instead canaliculi connecting lacunae to the Haversian canal calcify. This radial calcification destroys the original lamellar structure of an osteon, instead of stimulating a remodelling process. It was also speculated by Currey that in spite of the fact that remodelling is necessary to keep the mineral of the bone available for exchange with the body fluids, bone still maintains its structural function without undergoing a turnover without the live cells. Examples of such phenomena are observed in other mammals and therefore no matter how much we like to think of ourselves as being different, the possibility that the bone is dead before we are is a very real one.

Ductility of bone then does not depend so much on the age of bone as on the age of osteons. The largest ductility and energy absorption occurs in children's bones where there is not just the pull-out mechanism in operation but also some ductility in partially pulled-out osteons which bend and result in a green stick fracture. The same mechanism may also exist in an adult bone as is shown in Figure 22. The area illustrated on this photomicrograph is taken from the base of the Tattersal and Tappin specimen (Figure 18a). Therefore the radius of bending was very small and a pull-out osteon was also bent showing the mode of a green stick fracture. The remarkable fact about this mode of deformation is that the bone specimen belonged to a 57 year old man.

DISCUSSION

There is an obvious effect of morphology of bone on its fracture characteristics. In order to put this relationship in a better perspective, a small modification should be made in our minds to the traditionally accepted concept of bone structure. On the ultra-structural level, the concept that bone consists of an organic matrix with discontinuous crystallites of apatite dispersed throughout should be re-examined. With the development of electron scanning microscopy there is sufficient evidence that hydroxyapatite exists also in a continuous polycrystalline, generally lamellar morphology. Hookean behaviour of bone provides a further indication that such an assumption is more realistic.

Examining a basic unit of cortical bone - an osteon in more detail, its lamellar structure should be put under greater scrutiny. Assuming the traditional description that lamellae in osteon are bound together with a "cementing substance", the thickness of which is extremely small as

compared with the thickness of lamellae, appears to be also questionable. If that were the case, pull-outs of osteons having rough surfaces as shown in electron scanning photomicrographs would not be possible. Thus one should assume that the interlamellar substance is much thicker. Fractured osteons and deprotenized osteons show indeed quite wide spaces between the lamellae. On the basis of the work done in this laboratory the hypothesis is proposed that these spaces are occupied by an organic phase with discontinuous crystallites of apatite, as was earlier believed to be the case for the whole bone [2]. Such an aggregate of apatite and collagen would be very "ductile" easily deformable and would account for most of the ductility in bone. The presence of an organic-mineral aggregate which, under load, should display viscous or at least viscoelastic deformation would also account for the distinctive viscoelastic behaviour of bone. The continuous phase of hydroxyapatite, on the other hand, provides strength and stiffness to the cortex of long bones. A ductile-brittle transition as suggested by Hasson and Armstrong [53] indeed exists in bone but it may be more practical to look at it as a function of age rather than a function of temperature.

Another hypothesis is proposed describing in a little more detail the aging process in bone. Young osteons are made up of concentric cylinders consisting of collagen and the continuous polycrystalline phase of hydroxyapatite. Between the lamellae is an aggregate of collagen and apatite which, during cyclic loading, transmits hydrostatic pressure like a liquid. This process, according to a model proposed earlier, supplies nutrients to concentrically located bone cells. When the cells die, either remodelling occurs or radial calcification. Since the amount of remodelling decreases with age, old bones have more calcified osteons, and hence display a more brittle behaviour.

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TABLE 1

Strength and Elastic Modulus of Bone with Organic Phase Removed

Investigator	Tension		Compression	
	σ MPa	E MPa	σ MPa	E MPa
Sweeney [7]	7		37	7584
Mack [6]	6	17237	47	6688

TABLE 2

Some Data for Collagen in Tendon

Investigator	Density kg/m^3	Strength MPa	Elongation ε%	Elastic Modulus E MPa
Hall, 1951 [19]	10800		25	1480
Viidik, 1966 [22]	-	44-49	-	-
Piekarski, 1967 [20]	-	43	30	549

TABLE 3

Energy to Fracture of Cortical Bone

Investigators	Type of Bone	ε joules/m^2
Melvin and Evans, 1973 [33]	Fresh bovine	58,650
Piekarski, 1970 [3]	Fresh bovine	59,000
Saha, 1975 [34]	Embalmed human	20,700
Currey & Butler, 1975 [35]	Fresh human adult	12,000 - 18,000
	Fresh human children	16,000 - 22,000
Piekarski, 1976 [36]	Fresh human 78-85 yrs	23,000
	Fresh human 40-50 yrs	64,000

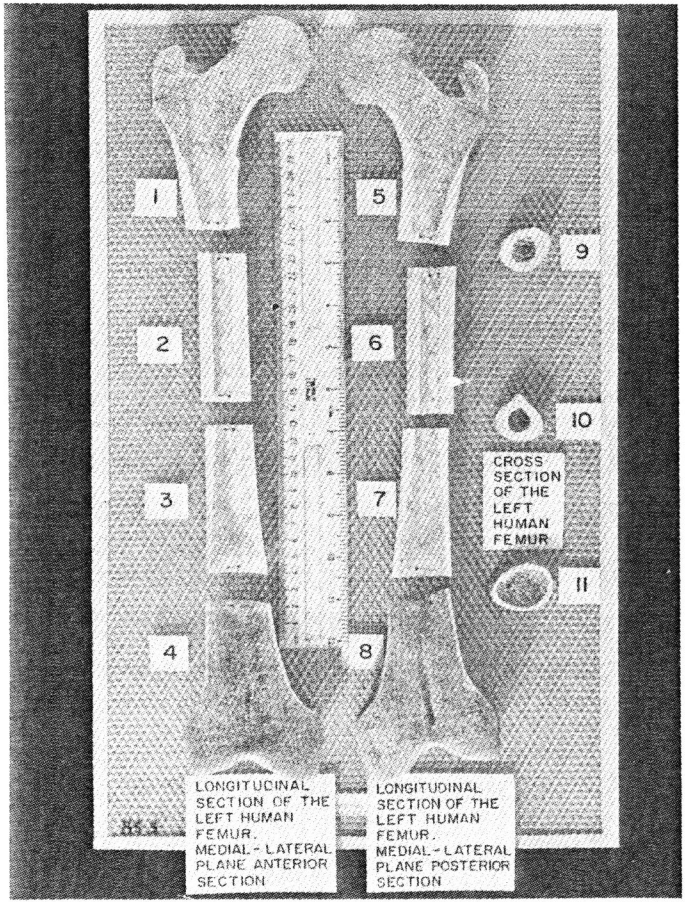

Figure 1 Distribution of solid material in a femur

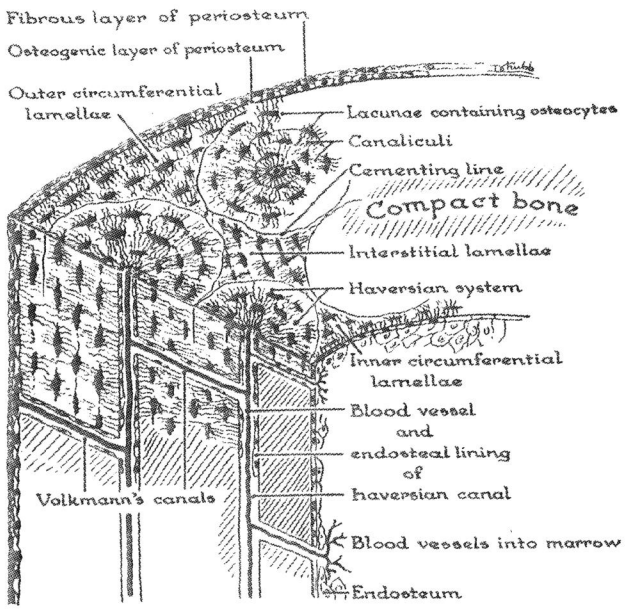

Figure 2 Schematic illustration of the microstructure of the cortical bone

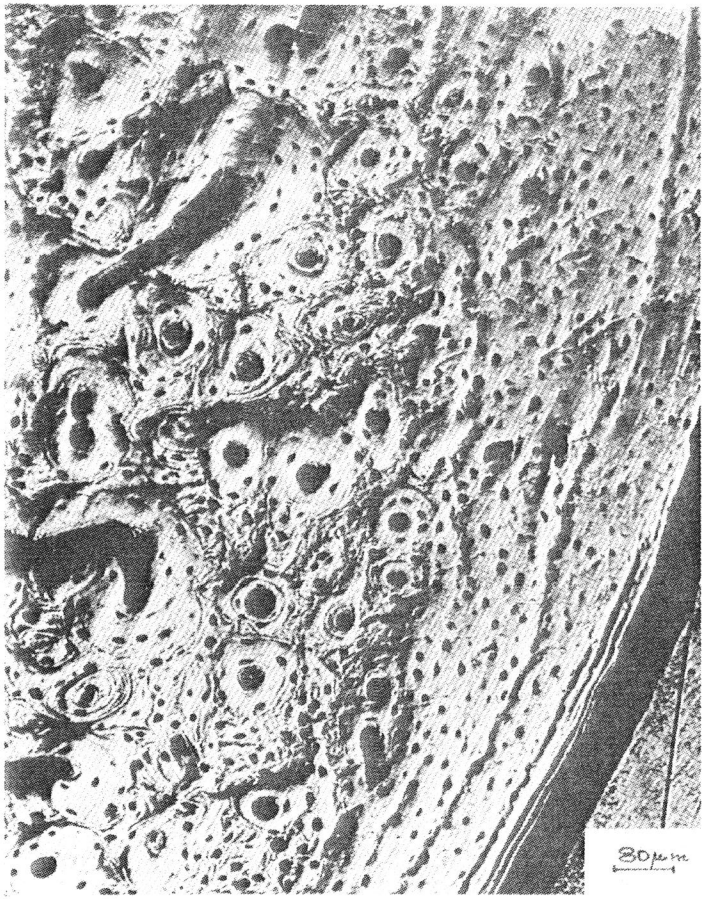

Figure 3 Cross section of the shaft of a cortical bone. Optical microscope

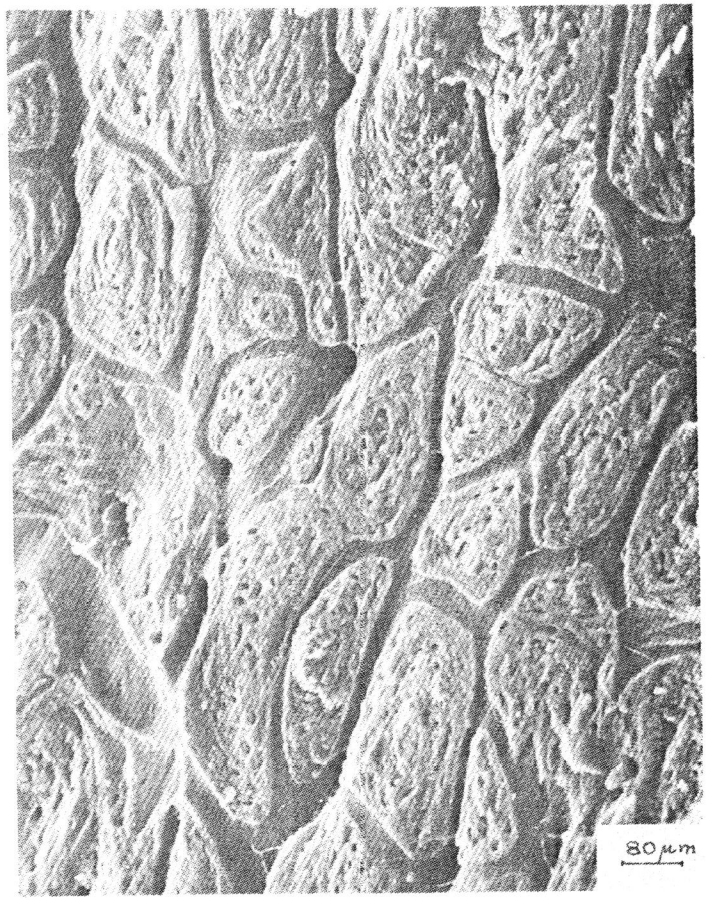

Figure 4 Longitudinally cleaved shaft of a cortical bone. Scanning electron microscope

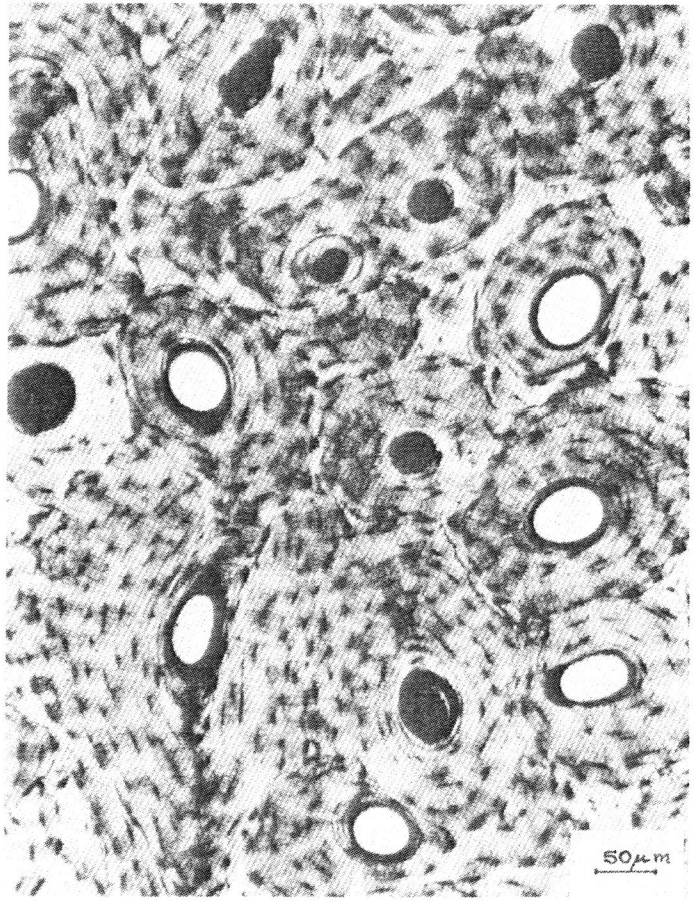

Figure 5 Thin section of a cortical bone examined with transmitted light on the optical microscope (Munro 1975).

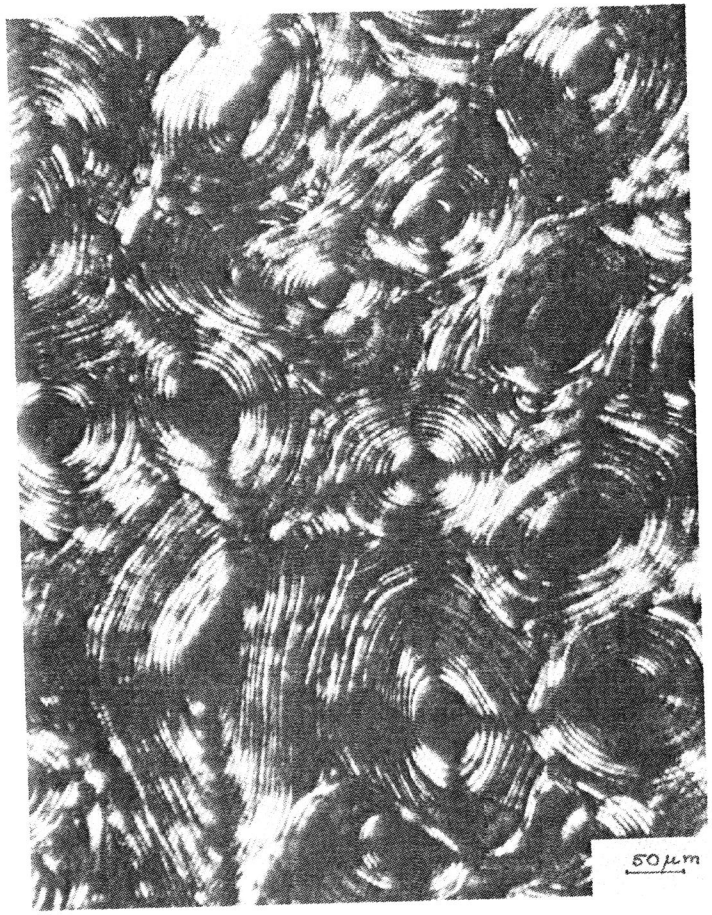

Figure 6 The same section of a cortical bone examined with polarized light on the optical microscope (Munro 1975).

Figure 7 Surface of an osteon etched with HCl acid viewed under scanning electron microscope. In spite of the expansion of the organic phase upon removal of the mineral, ridges standing up in relief indicate higher content of collagen in alternate lamellae.

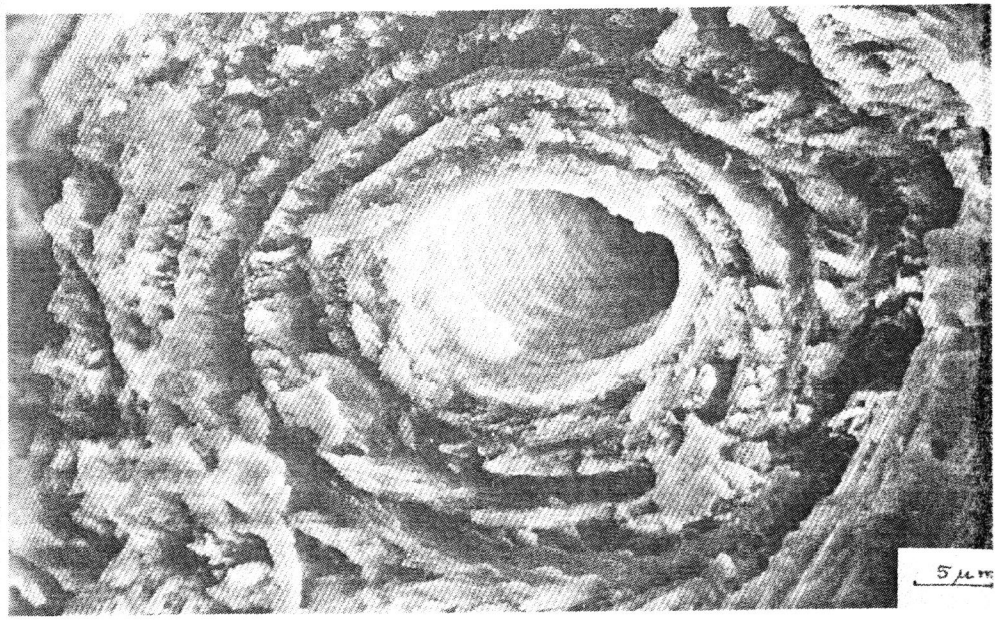

Figure 8 Osteon from the fractured surface of bovine bone treated with ethylene diamine. Structure shows polycrystalline concentric cylinders of hydroxyapatite

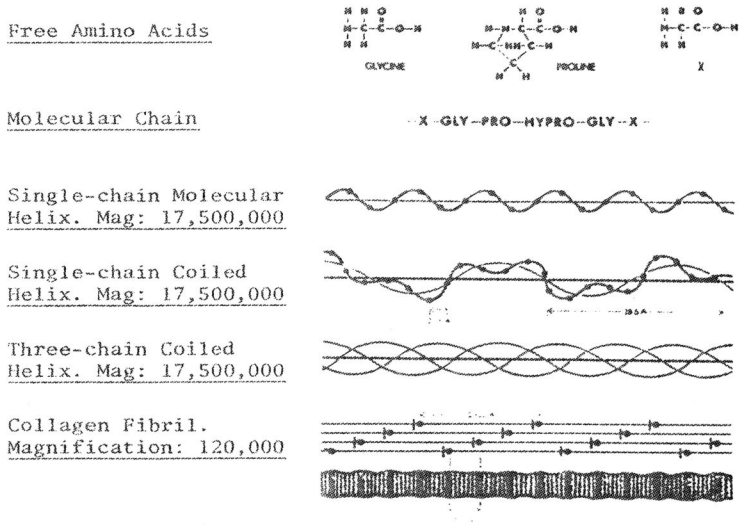

Figure 9 Schematic structure of collagen (after Gross 1961).

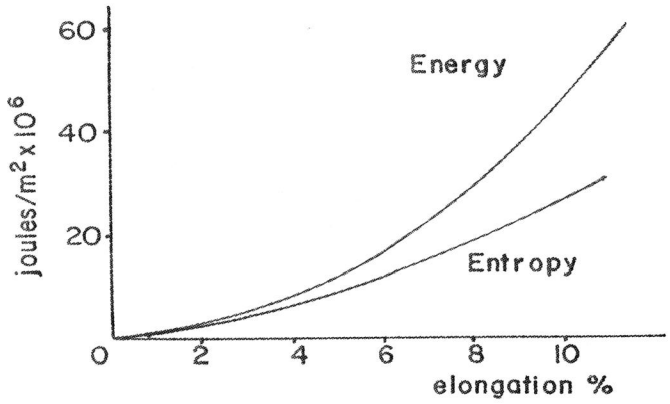

Figure 10 Energy and entropy contributions to the deformation of an animal tendon at pH 7.0 and 35°C (After Hall, 1952).

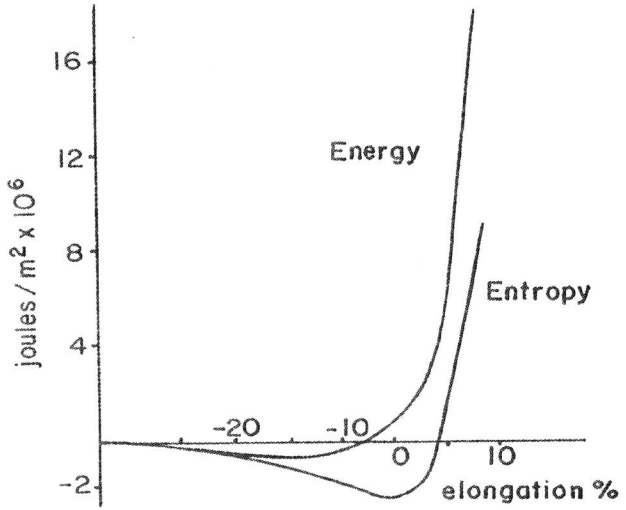

Figure 11 Energy and entropy contributions to the deformation of an animal tendon at pH 1.75 and 35°C (After Hall, 1952).

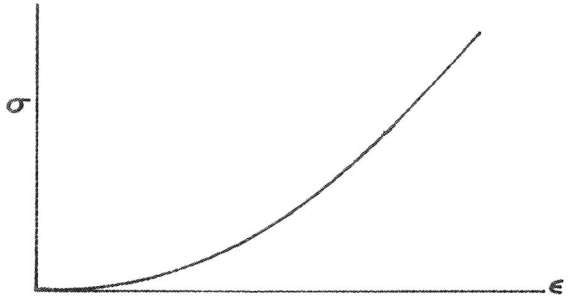

Figure 12 Typical stress-strain curve for a tendon

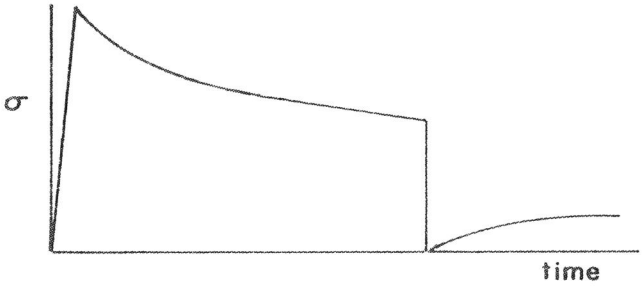

Figure 13 Typical relaxation - recovery curve for a tendon

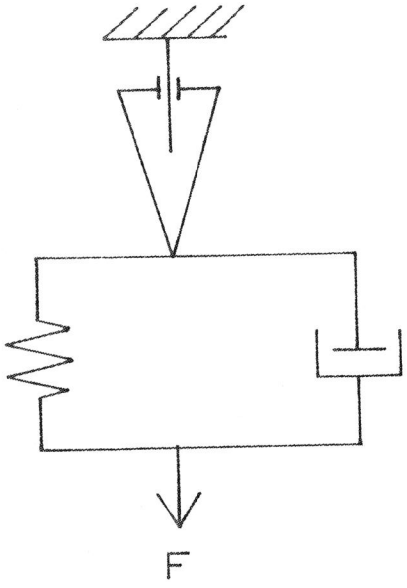

Figure 14 Rheological model of collagen fibers

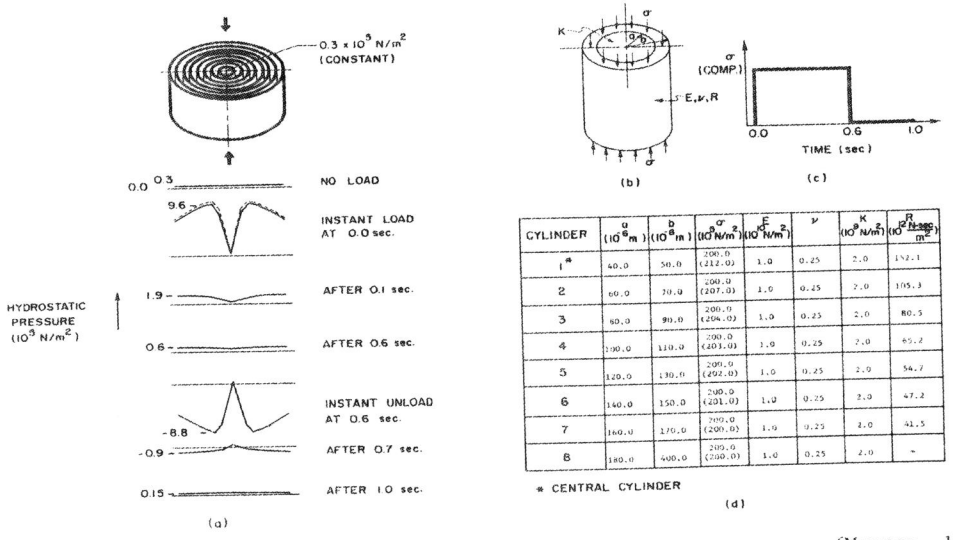

Figure 15 Model of an osteon subjected to compressive stresses (Munroe, 1976

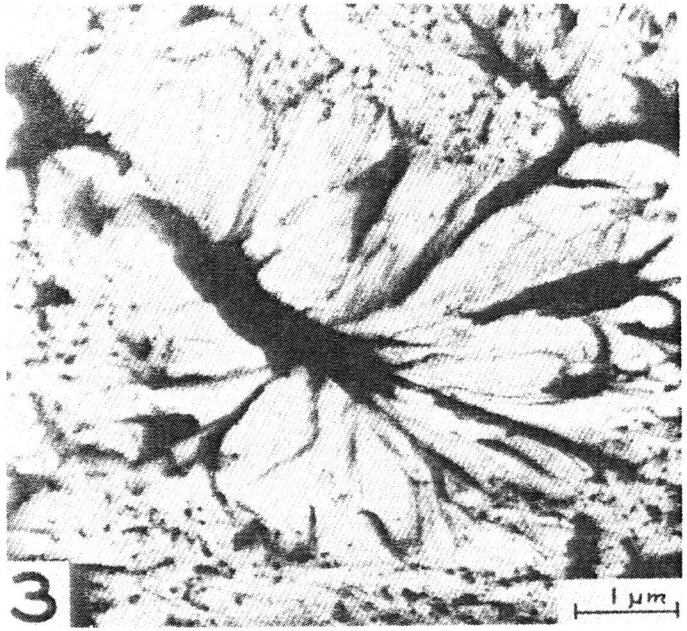

Figure 16 Electron microscope photomicrograph showing porosity of "the solid bone" and higher density protective sleeve around the canaliculus (After Voss, 1963).

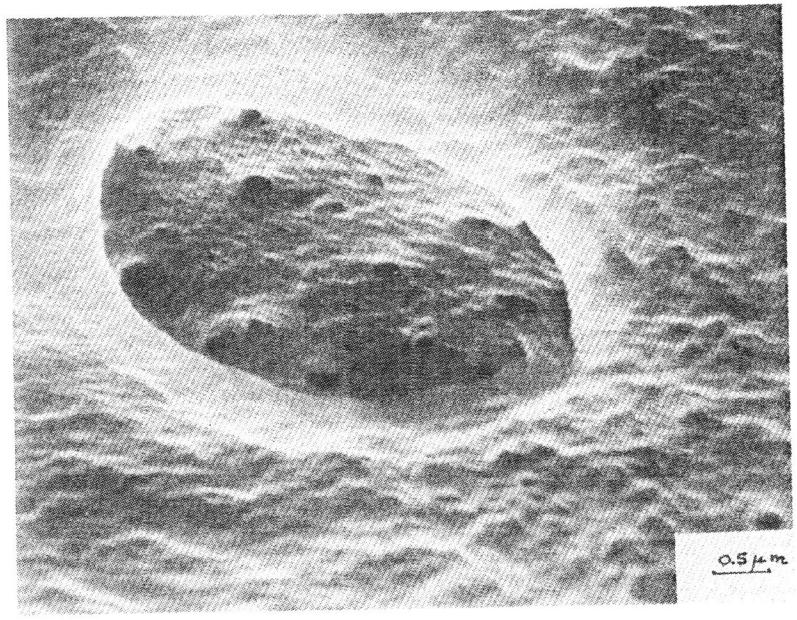

Figure 17 Scanning electron microscope photomicrograph of a lacunae. The organic phase has been removed with ethylene diamine showing dense mineral envelope acting as a protective interface for the lacunae.

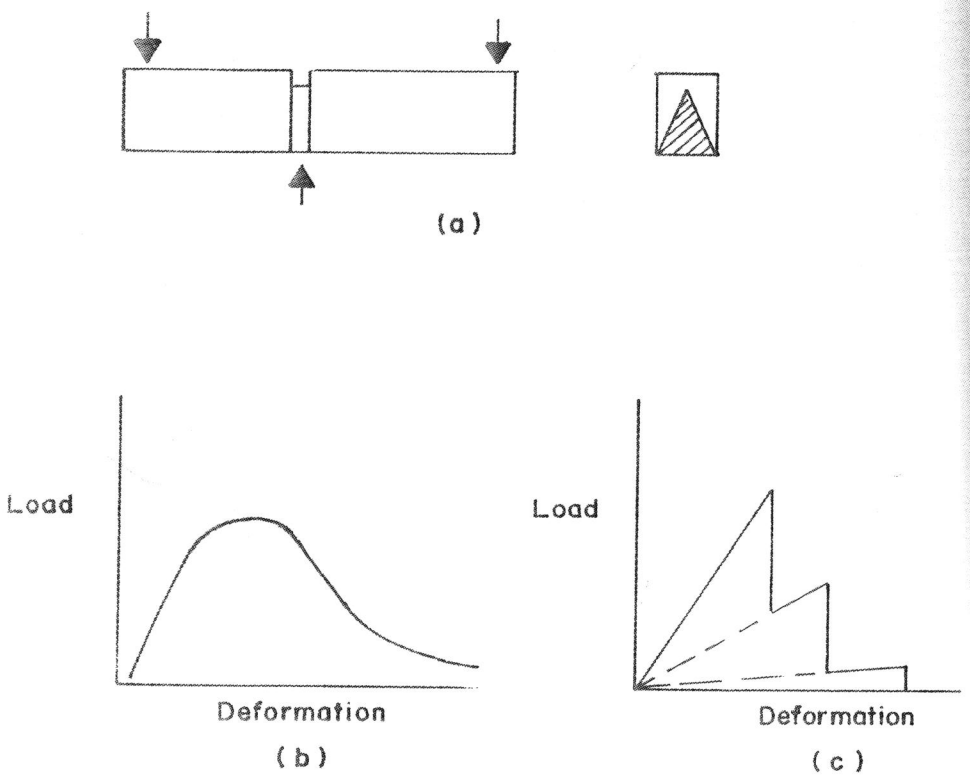

Figure 18 Typical load-deformation curves obtained in three point bending of Tattersal and Tappin Specimens.

(a) The shape of the specimen, (b) Typical curve of a ductile specimen, (c) Typical curve of a brittle specimen.

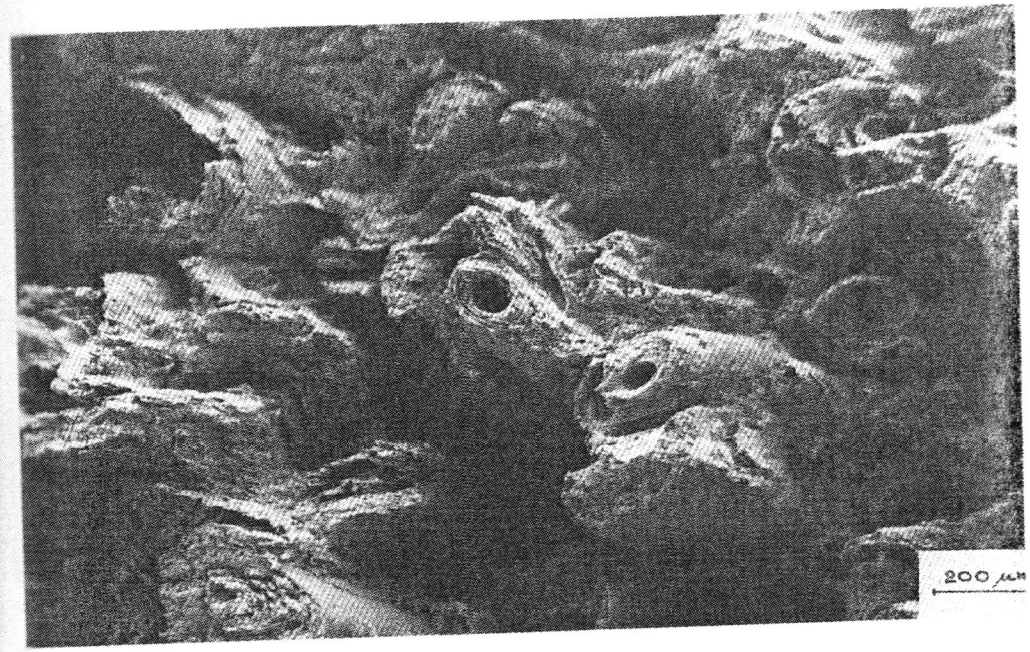

Figure 19 Fractured surface of a young human bone (23 years old) showing large portion of pulled-out osteons.

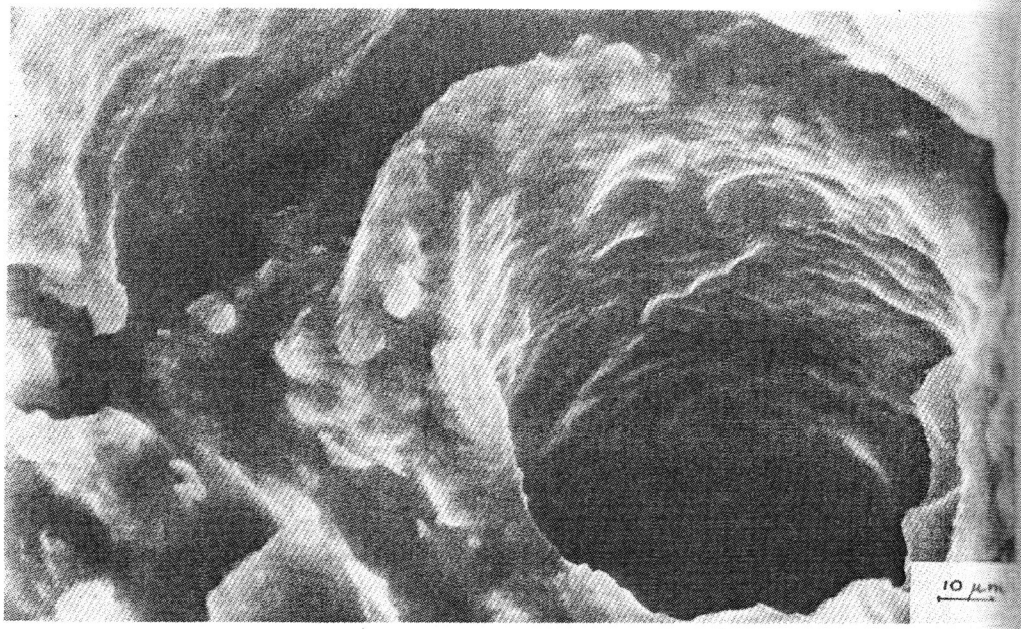

Figure 20 Structure of concentric cylinders of a young osteon which generally resulted in the "telescopic" pull-out.

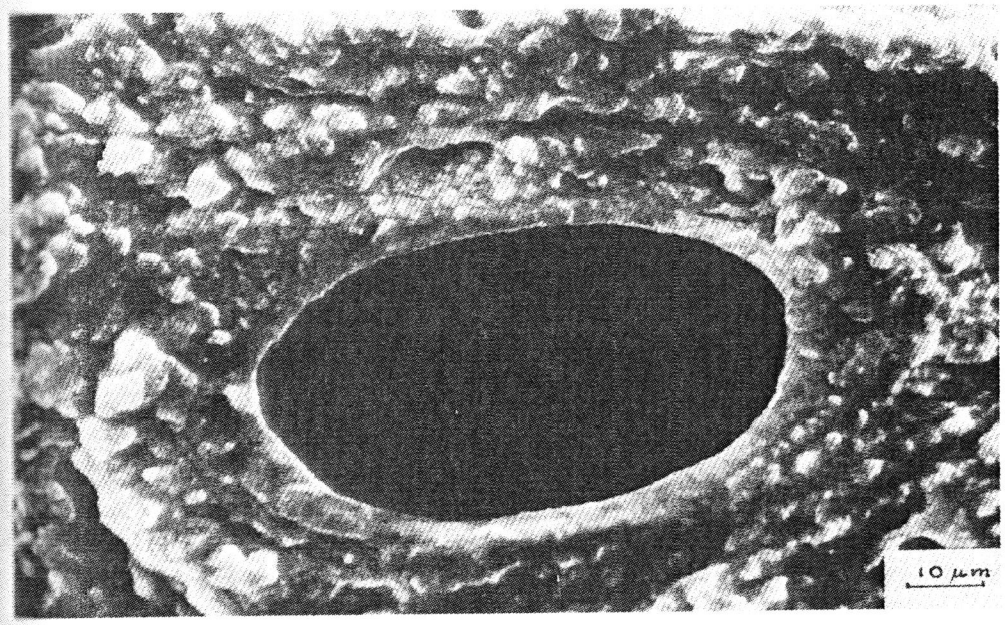

Figure 21 Structure of an old osteon indicating the previous existence of concentric rings which are presently bridged with calcified tissue. Human bone (83) years of age).

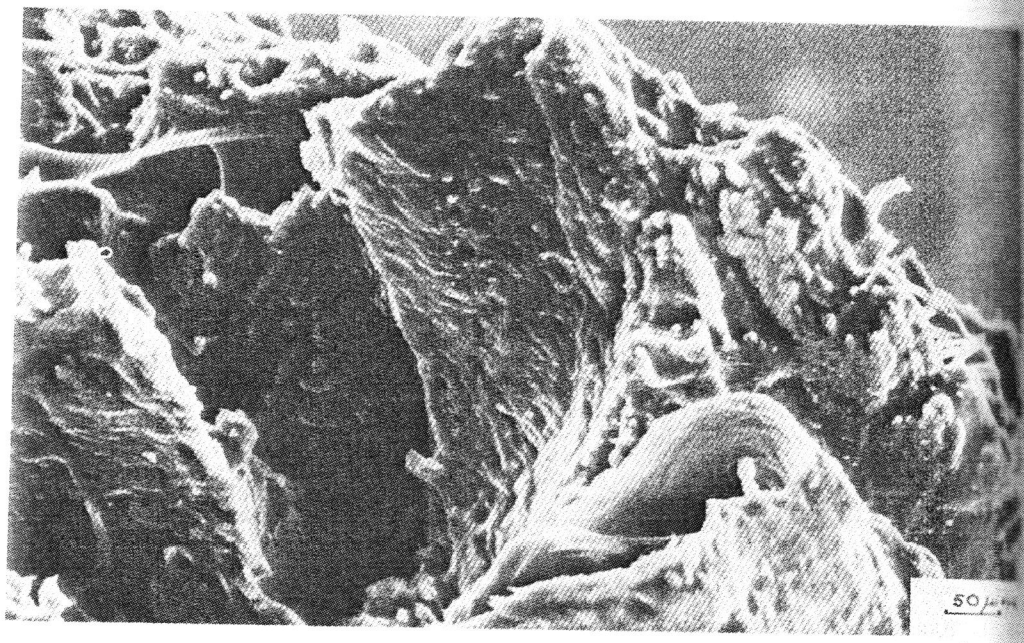

Figure 22 Young osteon in 57 year old bone. The pull-out occurred at the base of the specimen resulting in the curvature of an osteon bending over a small radius.