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Longitudinal shear properties of human compact bone and its constituents, and the associated failure mechanisms

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Human compact bone specimens were tested in longitudinal shear at two different strain rates. The maximum stress and energy absorption capacities were 50.40 ± 14.08 MN m^{-2} and 20720 ± 9310 J m^{-2} respectively for 14 embalmed specimens tested at a cross head speed of 2.1 × 10^{-6} m sec^{-1}. The maximum stress was found to be 75% of the transverse shearing strength. Bone specimens were also tested after selectively dissolving the organic and mineral components. The results showed that the composite strength of bone was much higher than the summation of strengths of its organic and mineral phases. Fractographic examination of the fracture surfaces showed that debonding of the interfaces between the osteons and the surrounding bone matrix and between the osteonal lamellae were the main mechanisms of longitudinal shear failure.

1. Introduction

In recent years, the mechanical properties of bone have been an area of active investigation by many authors and as a result, considerable quantities of data have been generated on the tolerance limits of bone [1–4]. A review of this literature clearly indicates that compared to the tensile and compressive properties of bone, only scanty information is available on its shearing properties. In particular, data on the longitudinal shearing strength of bone is lacking. Similarly, no data is also available on how the individual shearing strengths of the organic and mineral bone constituents contribute to the overall shearing strength of bone. Therefore the object of this investigation was to determine the longitudinal shearing properties of human compact bone. To relate the shear strength of bone to its organic and mineral phases, bone samples were tested after selectively dissolving one component. To further improve our understanding of the mechanisms of shear failure, the fracture surfaces were examined fractographically using scanning electron microscopy [5, 6]. Embalmed human bones were used in this study as fresh human bones were not easily available.

2. Previous investigations

Rauber [7] was the first author to report on the shearing strength of bone. He tested fresh compact bone specimens of 2 mm × 2 mm square cross section cut parallel to the long axis of the bone, loaded transversely (i.e. perpendicular to the direction of osteons), and obtained a shearing strength of 116.2 MN m^{-2}. Similar specimens, cut perpendicular to the long axis of the same femur and loaded parallel to the bone axis failed at less than half of the transverse shearing strength thus showing the directional differences in the shearing strength of compact bone.

Evans and Lebow [8] studied the regional differences in the transverse shearing strengths of embalmed femoral samples and found that wet samples from the middle (72.3 MN m^{-2}) and distal (64.5 MN m^{-2}) thirds showed the highest and lowest average shear strengths respectively. The shearing strength of the wet samples from the medial (70.3 MN m^{-2}) and posterior (70.1 MN m^{-2}) quadrants were approximately equal, while the anterior quadrant was the weakest (64.9 MN m^{-2}). The authors also found that air drying decreased the shearing strength by 18.3% (Table I) in contrast

TABLE I Comparison of mean longitudinal (loaded parallel to the long axis of bone or osteons) and transverse (loaded perpendicular to the long axis of bone) shearing strengths of human compact bone.

Source	Bone	Condition	No. of Specimens	Direction of Shear	Shearing Strength ($MN\,m^{-2}$)	Reference
30 year old man	Femur	Fresh	5	Transverse	116.2 (102.9–127.5)	Rauber [7]
	Femur	Fresh	5	Longitudinal	49.3 (41.7–63.7)	
5 white males of ages 58 to 81 years	Femur	Embalmed & wet	109	Transverse	67.6 (41.5–105.5)	Evans and Lebow [8]
	Femur	Embalmed & dry	109	Transverse	55.2 (24.7–87.0)	
Human (18 and 63 years old)	Femur	Fresh	3	Longitudinal	74.1 (63.7–86.3)	Frost *et al.* [9]
	Tibia	Fresh	5	Longitudinal	79.9 (62.8–93.2)	
Human	Femur	Wet		Transverse	82.4 ± 1.77	Ibuki [10]
	Tibia	Wet		Transverse	80.4 ± 1.86	
Human (age 25)	Femur	Wet	48	Longitudinal (single concentric osteons)	48.2 (38.2–55.9)	Ascenzi and Bonucci [11]

to tensile strength, which increased by 31.3% by drying. There was no decrease in the average shearing strength with advancing age.

By using a simple but rather primitive test apparatus, Frost *et al.* [9] investigated the longitudinal shear strength of bone by punching a 0.9 mm diameter needle through 70 μm thick cortical bone sections (Table I). They also tested 6 longitudinal bone specimens loaded perpendicular to the direction of osteons and obtained a mean transverse shearing strength of 82.4 (68.7 to 97.1) $MN\,m^{-2}$. This value is not included in Table I, as the authors themselves suggested that, due to the wrong method of applying shear, their result on the longitudinal sections should be discounted.

The only other data on macroscopic shearing strength of compact bone was reported by Ibuki [10] (Table I). Using compact bone bars of square cross section of 2.5 mm x 2.5 mm, the author observed that the ultimate shearing strength and ultimate slip in a direction parallel with bone axis corresponded to 70% and 85% of those in a direction perpendicular to bone axis for femoral samples of cattle, respectively. He also reported that the ultimate shearing strength of formalin-fixed compact bone corresponded to about 1.05 times and 1.1 times the fresh one for man and cattle, respectively, while the ultimate slip showed no difference between the two groups. The impact shearing strength was about half of the ultimate strength in normal test.

In contrast to the macroscopic shearing properties reported previously, the shearing properties of single osteons were studied by Ascenzi and Bonucci [11] (Table I). Their results showed that: (1) osteons having a marked longitudinal spiral course of fibre bundles in successive lamellae are least able to support shearing stress; (2) ultimate shearing strength and modulus of elasticity of osteons increase with increasing calcification; (3) the shearing strength of single osteons is markedly lower than the tensile and compressive strength for samples of the same type. They [11] also attempted to test the shear strength of osteons loaded eccentrically, but the irregularity of the eccentrically tested samples made it impossible to calculate stress as a function of the lateral surface.

3. Microstructure of bone

The microstructure of bone is well documented in the medical literature [12–15]. As shown in Fig. 1, the most characteristic feature of adult bone tissue is its lamellar structure, the collagen fibres and the calcified matrix being organized in thin layers of lamellae (4 to 12 μm thick). In most cases, the collagen fibres in each layer have a dominant direction which is different from those in adjacent layers and the lamellation is a result of this differing orientation of the collagen fibrils in each of the superimposed lamellae [16].

In general, the central shaft (diaphysis) of a long bone consists of several layers of inner and outer circumferential lamellae enclosing a middle

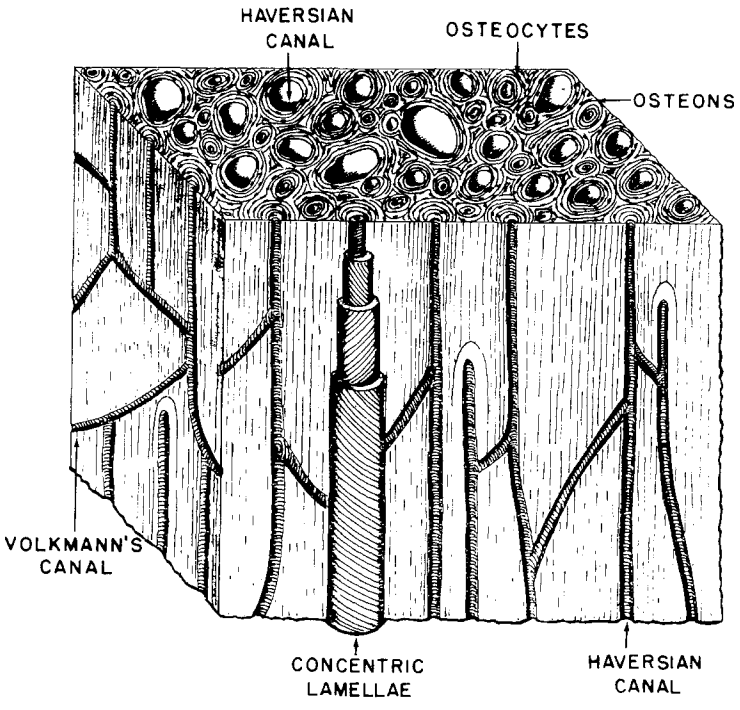

Figure 1 Microstructure of adult human compact bone (vertical axis is parallel to the long axis of the bone).

zone of longitudinally oriented osteons. An osteon can be defined as an irregular, branching and anastomosing cylinder composed of a more or less centrally placed neurovascular canal (5 to 100 μm diameter) surrounded by concentric cell-permeated lamellae of bone matrix [17], (Fig. 1). Osteons have often been described as the basic unit structure of compact bone [1, 12, 14, 18]. But this unit structure idea of osteon is often over-emphasized, and although a large part of adult human compact bone is composed of osteons, contrary to the popular belief, it is almost totally absent in bones of many species, e.g. white rats [13]. Compact beef-bone samples were also found to be predominantly non-osteonal in character [19]. Compact bone has several anastomosing vessel systems which carry nutrients to the bone cells (osteocyte). The most important amongst these is the central canals (Haversian canals, Fig. 1) of osteons, which communicate with each other and to the marrow cavity by transverse Volkmann's canals.

Chemically bone is generally described to be composed of three phases; organic matrix, bone mineral and water. The dry weight of bone is composed of 60 to 75% inorganic hydroxyapatite (both crystals and amorphous) and 25 to 35% organic matrix. Collagen accounts for as much as 90 to 97% of the organic matrix, the remainder being amorphous ground substance, mainly mucopolysaccharides. Individual collagen fibres are about 500 to 2000 Å thick, several microns in length. The hydroxyapatite crystals are deposited around the periphery of the collagen fibres with their *c* axis parallel to the long axis of the collagen fibrils. The exact nature of the bonding between the collagen and the hydroxyapatite is not known. Controversy also exists about the size and shape

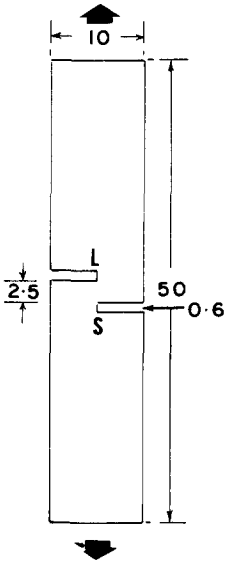

Figure 2 Approximate dimensions (in mm) of a longitudinal shear specimen, typical thickness being 1.5 mm. The width of the specimen is oriented in circumferential and thickness in radial direction. Applied directions of tensile loading to produce longitudinal shear failure of the face LS is shown by arrows.

of apatite crystals. It is generally believed that they are 200 to 400 Å long and 10 to 50 Å wide or in diameter, depending on whether or not the crystals are believed to be plates or needles.

4. Materials and methods

Longitudinally oriented compact bone specimens (Fig. 2) were machined from embalmed human femurs using a Bridgeport milling machine. Throughout the machining process, the specimen was constantly irrigated with water to minimize any possible thermal damage.

The specimens were divided into four groups and the first two groups of compact bone specimens were tested in longitudinal shear at two different strain rates. The other two groups of specimens were used to study the longitudinal shear properties of organic and mineral phases only.

The collagen was removed from the group III specimens by treating them with a 5% potassium hydroxide solution just below boiling point for 6 h. The collagen was continually skimmed off during this time. The specimens were then washed in running water overnight and kept in ethyl alcohol until testing 24 h later. In order to remove the mineral component, group IV specimens were treated under a vacuum with a 9% nitric acid solution for 24 h, and then rinsed in running water for 24 h. To test the effectiveness of demineralization, X-rays were taken of specimens treated with the acid and compared with a normal bone specimen as shown in Fig. 3. Results clearly showed removal of all mineral. In the subsequent portion of the paper, the following terminology will be used for the three groups of the specimens: (1) the first two untreated groups of specimens will be called normal or intact; (2) group III specimens will be denoted as deorganified, or mineral-only; (3) the group IV specimens will be denoted as demineralized or decalcified or protein-only, or collagen-only specimens.

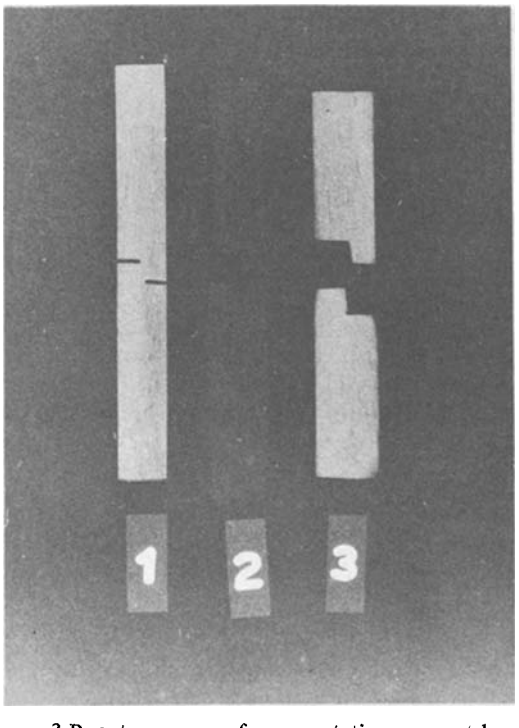

Figure 3 Roentgenogram of representative compact bone samples: (1) untreated bone (before failure); (2) demineralized containing organic component only (after failure); (3) deorganified containing mineral component only (after failure).

All specimens were mechanically tested in a floor model Instron testing machine using compressed air operated grips. The specimens were subjected to a tensile loading parallel to the long axis of the bone as indicated by arrows in Fig. 2. This caused longitudinal shear failure of the face LS. All samples were stored in normal saline until testing and they were tested in a wet condition. Load deformation curves were continuously recorded for all specimens.

After mechanical testing, fracture surfaces of selected normal bone specimens were defatted in acetone for improved resolution of natural bone cavities in the subsequent scanning electron microscope study. These specimens were coated with carbon and gold and examined in an ETEC Autoscan SEM. Fracture surfaces of specimens from which organic and mineral components were dissolved were coated with gold in a sputter coater and examined in an ISI Mini-SEM. Availability of the microscopes were the only reason for these two different treatments and equally satisfactory results were obtained in both cases.

5. Results and discussion

Altogether forty specimens were successfully tested including twenty-four normal, ten demineralized and six mineral only specimens. Typical stress deformation curves for these three groups of specimens are shown in Fig. 4. The stress deformation curve of normal compact bone in longitudinal shear (marked a in Fig. 4) is similar to the stress–strain curves of single osteons in shear, reported by Ascenzi and Bonucci [11]. Fig. 4 shows that the stress deformation behaviour of normal bone

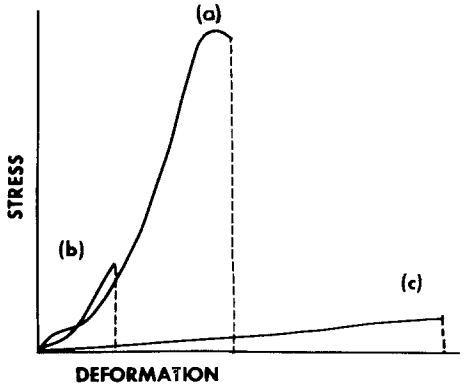

Figure 4 Typical stress deformation curves of: (a) intact bone specimens, (b) specimens with mineral component only, (c) specimens with organic component only (decalcified).

in longitudinal shear is nonlinear and there is a yield point, although the specimen fractures soon after yielding. On the other hand the stress deformation behaviour of decalcified specimens (marked c in Fig. 4) was linear and did not show any yielding. This agrees with similar tensile behaviour of fully decalcified specimens observed by Burstein *et al.* [20].

Fig. 4 also shows that when mineral component alone is tested in longitudinal shear, the specimen fails in a characteristically brittle fashion with minimum amount of deformation compared to normal bone. This indicates that loss of collagen also causes loss of ductility of the bone specimens. Similarly comparing curves a and c in Fig. 4, we find that when collagen alone is tested in longitudinal shear, it can support only a small amount of stress but can sustain a large amount of deformation. This supports the hypothesis that it is the organic component of bone which is mainly responsible for its ductility.

The ultimate longitudinal shear stress was calculated from the maximum load divided by the surface area of the face LS. The energy absorption capacity of the specimens was obtained by measuring the area under the load deformation curves by a planimeter.

The means and standard deviations of the longitudinal shear properties calculated from the load–deformation curves are presented in Table II. The maximum longitudinal shear stress of normal bone specimens agreed with the punching shear strength of single osteons measured by Ascenzi and Bonucci [11]. The maximum longitudinal shear stress at slow strain rate (50.4 MN m^{-2}, Table II) was approximately 75% of the transverse shearing strength of wet embalmed femoral specimens obtained by Evans and Lebow [8]. This points out the anisotropic character of the mechanical properties of bone which is now well recognized [21]. Rauber [7] also found that average shearing strength of compact bone was considerably less when loaded parallel to the long axis of the bone than when loaded perpendicular to the long axis (Table I).

Table II also shows that when the cross-head speed is increased from 2.1 to 21.1 $\mu m\,sec^{-1}$, the maximum stress is reduced from 50.4 to 42.9 MN m^{-2}; however the change in the energy absorption capacity is small. In a subsequent study it was found that longitudinal shear strength of compact bone remained approximately constant up to a cross-head speed of about 211 $\mu m\,sec^{-1}$, beyond which it reduced drastically [22]. This is similar to Bonfield and Datta's [23] finding that the transverse tensile strength of compact bone under shock loading condition was considerably less than the tensile fracture stress of transversely oriented specimens tested quasistatically.

When the maximum longitudinal shear stress and energy absorption capacity of collagen-only (decalcified) and mineral-only (collagen dissolved) specimens are compared with the strength of intact bone, we found that the mineral component alone could contribute about 24% of the maximum stress and 14% of the energy absorption capacity of normal bone. On the other hand the organic component had only 8% of the maximum stress but 27% of the energy absorption capacity of normal bone. Thus these two different phases with contrasting mechanical properties, when combined, achieved a composite strength which was much higher than the summation of their individual strengths. It should be pointed out that during the

TABLE II Mechanical properties (mean ± one s.d.) of embalmed human compact bone specimens in longitudinal shear

Condition	No. of Samples	Crosshead Speed ($m\,sec^{-1}$)	Maximum Stress ($MN\,m^{-2}$)	Energy Absorption Capacity ($J\,m^{-2}$)
Embalmed	10	21.1×10^{-6}	42.86 ± 9.05	19,670 ± 13460
Embalmed	14	2.1×10^{-6}	50.40 ± 14.08	20,720 ± 9310
Decalcified	10	2.1×10^{-6}	4.18 ± 1.09	5,668 ± 2295
Collagen Dissolved	6	2.1×10^{-6}	12.14 ± 2.40	2,990 ± 1586

decalcification and deproteinization procedures, the structural integrity of the organic and mineral phases might have been affected somewhat; therefore their *in vivo* strengths might be slightly higher than the values reported here. Embalming also might have caused some changes in the mechanical properties, however, it has been shown that its effect on tensile properties was minimal [24].

Unlike mechanical properties of normal bone, only a few studies have been conducted on the properties of the organic and mineral components of bone, and therefore it is difficult to compare the results of this study with the data observed by others. However, similar results have been reported by Mack [25] who tested the individual components of bovine bone in tension and compression. In his study the mineral component had 30% of the strength of corresponding intact bone in compression and 5% in tension. Similarly the protein had less than 0.1% strength in compression and 7% in tension. Ko [26] also reported that the ultimate tensile strength of decalcified bone was

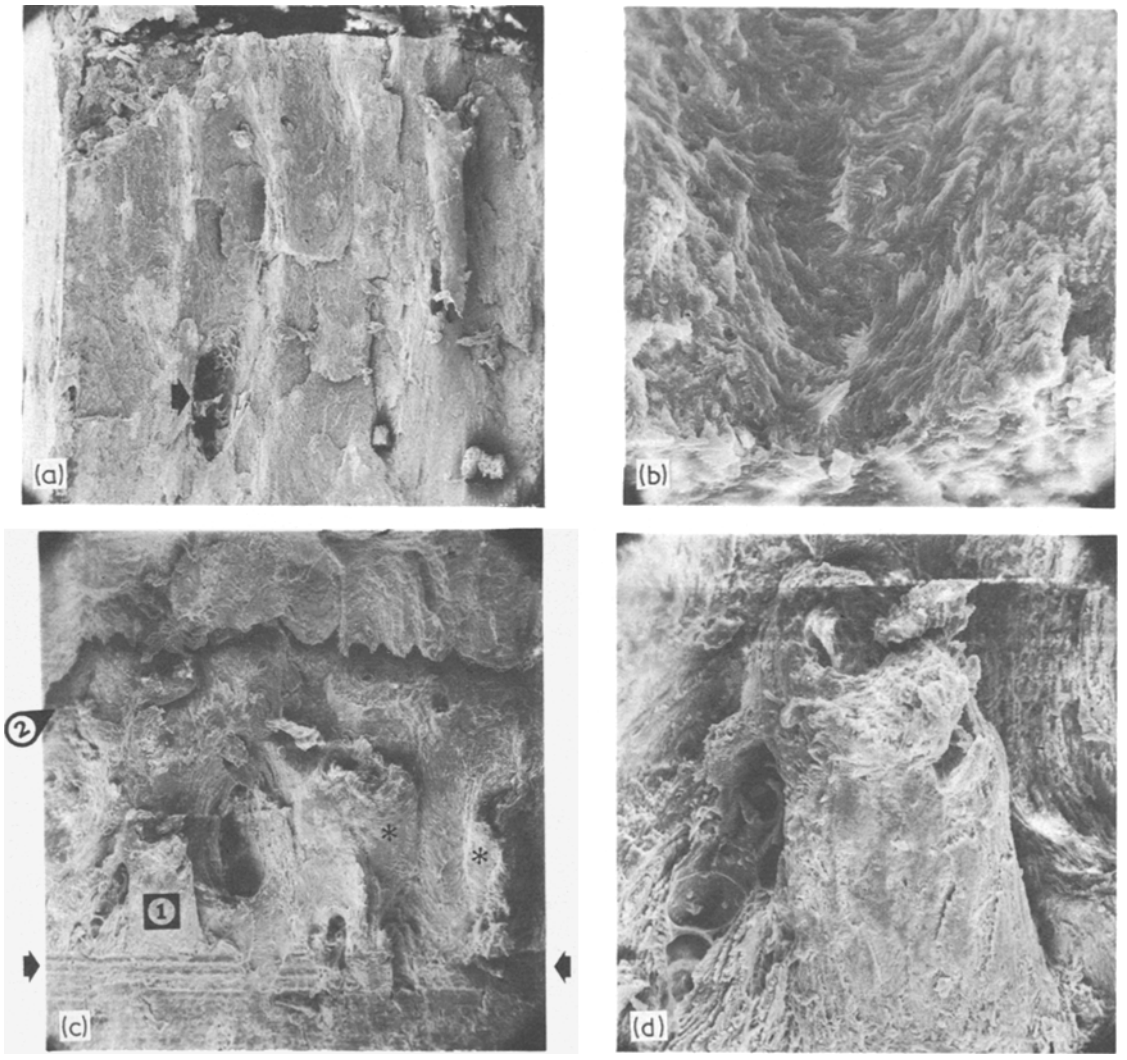

Figure 5 Scanning electron micrographs of longitudinal shear fracture surfaces of human compact bone specimens. (a) Vertical channel-like depressions caused by pull out of osteons. The arrow shows the opening of a transverse Volkmann's Canal (original magnification 50X). (b) The hollow left by a pulled out osteon showing longitudinal shear failure along osteonal lamellae (original magnification 400X). (c) The horizontal arrows indicate the edge of the notch or beginning of the shear fracture surface. Delamination or pull out of the osteons is marked by 1 (on the left) and by stars (on the right). A secondary microcrack is shown by 2. (original magnification 40X). (d) A magnified view of the pulled out osteon, marked 1 in Fig. 5c, showing failure along the cement line (original magnification 200X). Reduced in reproduction to 75% original size.

14% of that of normal undecalcified bone.

Fracture surfaces of most normal bone samples did not coincide with the face LS (Fig. 1), but were somewhat inclined to it. This may be due to the fact that notches in the specimens might have caused local stress concentration effect in the notch root. Moreover, the face LS was also subjected to a slight bending moment producing small normal stresses which accompanied the predominant longitudinal shear stress.

To gain further insight into the micromechanisms of longitudinal shear fracturing process, fractographic investigation of selected bone samples was conducted. Macroscopic appearances of most fracture surfaces exhibited a fairly rough texture. Further examination of the fracture surface by SEM indicated that debonding of the interface between the osteonal lamellae or delamination of the cement lines was the main mechanism of longitudinal shear failure. This is evident from Fig. 5 which shows that failure surfaces coincided with the interfaces between adjacent lamellae or followed the cement lines surrounding secondary osteons. Perhaps the main reason for such a failure is that few collagen fibres cross from one lamella to the other causing the potential weakness of the interface. Other authors have also shown this obvious preference for the crack to propagate between the lamellae or along the cement lines for certain failure modes, indicating the existence of a weak interface [27–33]. This may also be the reason for the significantly less transverse tensile strength of bone compared to its longitudinal strength [27]. However there may be other beneficial effects of such a weak interface such as they might act as a crack-stopping mechanism, and thus may increase the fracture toughness [5–6, 30–34].

This dual role of a weak interface also exists in engineering composite materials. For instance, for unidirectional carbon–carbon composite, it has been shown that the transverse tensile and longitudinal shear strength is quite low compared to longitudinal tensile strength [35] (similar to bone). On the other hand it has also been demonstrated that control of cracks by interfaces is possible in composites and thus "the toughness of a composite material having periodic stiffness-change along the crack path, may be very much greater than the toughness of the individual components of the composite" [36].

Although the fracture surface often followed

Figure 6 Scanning electron micrograph of a longitudinal shear fracture surface showing fracture through an osteon. The Haversian canal of the osteon is shown at the middle of the picture, vertically oriented (original magnification 200×). Reduced in reproduction to 75% original size.

the cement lines, other mechanisms also contributed to the longitudinal shear failure. Fig. 6 shows such an example where the failure occurred by shearing though an osteon, almost dividing it into two halves.

Fig. 5c shows that occasionally secondary microcracks (marked 2) also accompanied the main fracture surface. Such secondary cracks absorb energy by creating additional fracture surfaces and thus may contribute in increasing energy absorption capacity of the specimen [5, 31].

Fig. 7 shows typical alternate arrangement of collagen fibres in successive lamellae on a longitudinal shear fracture surface. This points out that fractographic examination of bone fracture surface by SEM is not only helpful for correlating surface features with fracture mechanisms, but it may also provide additional valuable information about the microstructural aspects of bone. When the fracture surface was examined at a hgiher magnification, it showed layered structure of mineral crystals associated with the collagen layers as shown in Fig. 8. Fig. 8 also shows that at the ultrastructural level, surface features are similar to the characteristics of somewhat ductile fracture.

While the fracture profile of most normal bone samples were inclined to the face LS (Fig. 2), fracture surfaces of decalcified and mineral-only specimens coincided with the face LS (Fig. 3). Fractographic examination of these decalcified

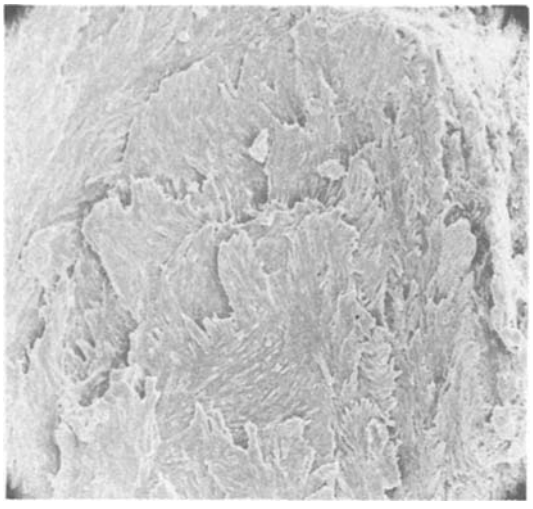

Figure 7 Scanning electron micrograph of a longitudinal shear fracture surface showing alternating arrangement of collagen fibres in successive osteonal lamellae (original magnification 400×). Reduced in reproduction to 75% original size.

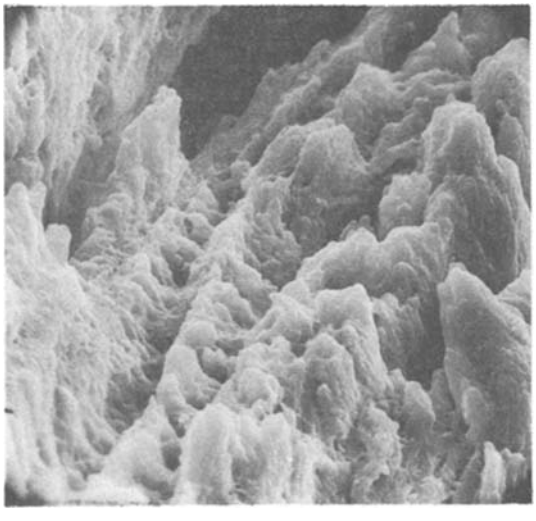

Figure 8 Scanning electron micrograph of a longitudinal shear fracture surface at high magnification showing layered structure of mineral crystals associated with collagen fibres. Signs of small dimples characteristic of ductile fracture are also visible (original magnification 5000×). Reduced in reproduction to 75% original size.

(protein-only) and mineral-only samples showed surface features which were qualitatively similar to the intact bone. This is exemplified in Fig. 9 which shows vertical channel like depressions produced by sliding out of osteons and these are similar to the surface morphology shown in Fig. 5a. However, at higher magnification, fracture surfaces of these specimens were smoother compared to intact bone.

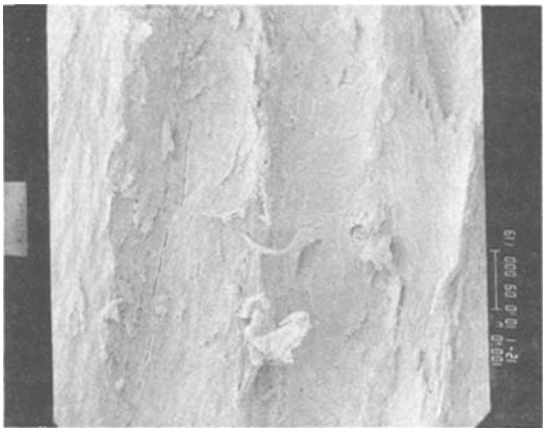

Figure 9 Longitudinal shear fracture surface of a mineral-only specimen (organic component dissolved) showing the vertical channel like depressions due to pull-out of the osteons. The vertical marker at right indicates a length of 100 μm (scanning electron micrograph, original magnification 120×). Reduced in reproduction to 63% original size.

This agrees with the observed mechanical properties as the fracture energy of the decalcified and deorganified samples were significantly lower than that of normal bone (Table II).

6. Conclusions

(1) Standardized human compact bone specimens were tested in longitudinal shear. The maximum stress and the energy absorption capacity were $50.4 \pm 14.1\ \mathrm{MN\,m^{-2}}$ and $20\,720 \pm 9310\ \mathrm{J\,m^{-2}}$ respectively for 14 embalmed samples (crosshead speed $2.1 \times 10^{-6}\ \mathrm{m\,sec^{-1}}$).

(2) The stress deformation curve of compact bone in longitudinal shear was nonlinear, and followed a slightly S-shaped curve. The mineral-only specimens (protein dissolved) failed in a brittle fashion with minimum strain while the organic component (decalcified) behaved elastically and failed only after undergoing significant amount of strain. The results indicate that it is the organic component which is mainly responsible for the ductile behaviour of bone.

(3) Mineral component alone could support only about 24% of the maximum stress and 14% of the energy absorption capacity of intact bone. On the other hand, organic component of bone could sustain about 8% of the maximum stress but 27% of the energy absorption capacity of normal bone. Thus the strength of normal bone was vastly superior than the combined strengths of its organic and inorganic components.

(4) The results of fractographic examination indicated that delamination of the interfaces between osteonal lamellae and debonding of the cement lines surrounding secondary osteons were important mechanisms of longitudinal shear failure.

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