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Progressive post-yield behavior of human cortical bone in shear

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Abstract

Bone fragility depends on its post-yield behavior since most of energy dissipation in bone occurs during the post-yield deformation. Previous studies have investigated the progressive changes in the post-yield behavior of human cortical bone in tension and compression using a novel progressive loading scheme. However, little is known regarding the progressive changes in the post-yield behavior of bone in shear. The objective of this short study was to address this issue by testing bone specimens in an inclined double notch shear configuration using the progressive loading protocol. The results of this study indicated that the shear modulus of bone decreased with respect to the applied strain, and the rate of degradation was about 50% less than those previously observed in compression and tension tests. In addition, a quasi-linear relationship between the plastic and applied strains was observed in shear mode, which is similar to those previously reported in tension and compression tests. However, the viscous responses of bone (*i.e.* relaxation time constants and stress magnitude) demonstrated slight differences in shear compared with those observed in tension and compression tests. Nonetheless, the results of this study suggest that the intrinsic mechanism of plastic deformation of human cortical bone may be independent of loading modes.

Keywords

Cortical bone; shear; post-yield; plastic deformation; viscous behavior

1. INTRODUCTION

Studying post-yield behavior of bone is essential to understand the different pathways of energy dissipation in the tissue [1, 2]. In previous studies, we have utilized a novel progressive loading protocol to study progressive changes in the post-yield behavior of human cortical bone in tension [3–5] and in compression [6, 7]. It was observed that the post-yield behavior of human cortical bone exhibited similar trends in both tension and compression tests, showing an exponential decay of elastic modulus, quasi-linear increase in plastic deformation, and bimodal viscous responses with increasing applied strain.

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The post-yield behavior of human cortical bone in shear has been reported in torsional tests [8–11]. Since the torsional tests generate non-uniform shear in the specimens [9–11], they are not suitable for studying bone tissue behavior in pure shear mode [11]. To address the issue, several pure shear tests, such as Iosipescu, Arcan, and double notch shear tests, were employed to determine the monotonic shear properties, such as shear modulus and strength of human cortical bone [8, 11, 12]. However, little is known regarding the progressive changes in the post-yield behavior of bone under pure shear condition.

In this short study, we extended the progressive loading approach to examine the post-yield behavior of human cortical bone in pure shear using an inclined double notch shear test. Such information is important in understanding age-related bone fragility and establishing the constitutive model for the finite element analysis of human cortical bone.

2. MATERIALS AND METHODS

2.1 Specimen preparation

Bone specimens were obtained from six human cadaveric tibias (age: 83.7 ± 5.1 years old), including three females (84, 88 and 88 years old) and three males (76, 79 and 87 years old) through the Willd Body Program (The University of Texas Southwestern Medical Center at Dallas, TX). All donors were screened for any known bone disorders (*e.g.* osteoporosis, osteomalacia, and osteopetrosis). Rectangular beams with dimensions of 2.0 mm $\times$ 4.0 mm $\times$ 50 mm were prepared longitudinally from the anterior aspect in the middle shaft of each tibia using a high-speed diamond saw under continuous irrigations of a PBS solution (Fig. 1a). Then, two notches of 0.5 mm wide and 2.0 mm deep were produced in the opposite side of the central region of the specimens using a diamond saw with a distance of 4.0 mm apart to ensure a shear loading in the gage region (Fig. 1b). Bone specimens were wrapped in PBS soaked gauze and stored in a freezer at -25°C prior to testing. Before testing, metal foil shear strain gages (Gage Series 062DY, Micro-Measurements, Vishay Precision Group, Raleigh, NC) with a gage length of 1.6 mm and a gage resistance of 120 Ω were glued between the two notches within the central (gage) region of the specimens (Fig. 1b). To protect strain gage region from moisture induced by irrigation during the mechanical testing, wax coating was applied to seal the strain gage and wiring junctions from outside environment.

2.2 Inclined double notch shear testing

Double notch shear tests have been widely used to measure shear properties of composite materials [13] and human cortical bone [8]. However, due to the localized high stresses at the notch roots of the specimens, the deformation and failure of the specimens are caused not only by shear, but also other factors, such as bending [14]. In this study, an inclined double notch shear test was employed to sustain a pure shear mode in the gage region (Fig. 1c) by removing the bending stress components at the notches [15, 16]. For different notch distances with a given specimen thickness ($b = 4$ mm in this study), a different tilting angle is needed. For example, the tilting angle is 33.5° for a relative notch distance of $L/b = 1$ selected in this study. Detailed description can be found elsewhere in the literature [15]. The average shear stress for an inclined double-notch shear test is calculated as

$$\tau = \frac{F \cos \alpha}{(L - w)W} \quad (1)$$

where F is the external force applied by the mechanical testing machine and measured in a load cell; L is the distance between the centers of two notches on the bone specimen; w is

the notch width; W is the width of the bone specimen; α is the tilting angle of the inclined double notch shear test.

Measurement of shear strain (γ) was implemented by connecting shear strain gages to a signal conditioning amplifier (Model 2100 system, Micro-Measurements, Vishay Precision Group, Raleigh, NC). Calibration of strain gages was performed with a shunt calibration unit built inside the signal conditioning amplifier. Output from the signal conditioning amplifier was sent to the control computer of the mechanical testing system for recording of shear strain (γ) during the test.

A pair of supporting jigs with a tilting angle ($\alpha = 33.5^\circ$) were manufactured to conduct the inclined double notch shear test (Fig. 1c). Initially, rubber bands were used to keep the two jigs together and hold the supporting pins in place. Bone specimens were then inserted inside the shear testing fixture and adjusted to a stable position. The inclined double notch shear testing jigs with the mounted bone specimens were placed between a pair of compression platens in the loading axis of the mechanical testing machine. The rubber bands were removed after a preload had been applied to stabilize the position of the whole test assembly.

2.3 Progressive loading protocol

Progressive loading protocol was conducted on human cortical bone specimens in shear mode in an electromagnetic mechanical testing system (Bose ELF 3300, Bose Corporation, Minnetonka, MN) following a well established procedure [3, 5, 6, 7]. Briefly, specimens were tested in a series of load-stress relaxation dwell-unload-viscoelastic creep dwell-reload (diagnostic) cycles with an incremental cross-head displacement (Fig. 1c) [6, 7]. After a preload of 10 N in compression, each specimen was loaded with a series of loading cycles at incremental strains (displacements). In each cycle, the specimens were loaded under the displacement control with a rate of 0.005 mm/s, held at the displacement level for 120 seconds, unloaded to 25 N, and held at the 25 N for 120 seconds. The dwelling time (120 s) was determined through pilot studies to ensure that the specimens reach to a quasi-equilibrium condition. The specimens were loaded in cycles with an incremental loading until failure. The incremental displacement levels were selected as 0.20 mm, 0.25 mm, 0.30 mm, 0.35 mm, 0.40 mm, 0.45 mm, 0.50 mm, 0.55 mm, 0.60 mm, 1.50 mm in this study. The shear stress (τ) was calculated from the load recorded from the force transducer (see Eq. 1) and shear strain (γ) was obtained from recordings of strain gages, respectively. The whole test was usually completed in 8 to 10 cycles (Fig. 2a). Due to the limited number of loading cycles, the effect of fatigue was ignored in this study. In addition, our pilot study showed that viscoplastic creep plays a negligible role in bone deformation and failure during the progressive loading test.

2.4 Mechanical testing calculations

Elastic, plastic and viscous behaviors of cortical bone specimens were assessed at each progressive loading level (Fig. 2) with a custom MATLAB program. The initial shear modulus (G_0) was calculated as the slope of the linear portion of the loading curve in the first cycle. The elastic modulus (G_i) at each applied strain was calculated as the slope of a line between the points at the end of stress relaxation dwelling and creep dwelling. The applied strain (γ_i) at each cycle was defined as the strain at the beginning of stress relaxation. For ease of comparison, an exponential decay of shear modulus (G_i) was assumed with respect to applied shear strain (γ_i) during the progressive loading as seen in tension and compression tests [3, 5–7]. The following equation was used for curve fitting to express the relationship:

$$G_i = G_0 e^{-m\gamma_i} \quad (2)$$

where, m could be used to estimate the sensitivity of bone to damage accumulation.

A linear regression equation was used to represent the relationship between applied strain (γ_i) and plastic strain (γ_p).

$$\gamma_p = A\gamma_i - B \quad (3)$$

The plastic strain (γ_p) was defined as the residual strain at the end of creep dwell. The yield strain ($\gamma_y = B/A$) was determined from the plot of applied strain vs. plastic strain. The yield stress (τ_y) was determined as the stress at the yield strain (γ_y) based on the applied stress vs. applied strain curve.

The progressive change in viscoelastic properties of bone in shear was determined in stress relaxation dwells. The stress vs. time data were extracted from each cycle and a built-in MATLAB function for nonlinear regression was applied to fit the data with the following equation:

$$\tau = \Delta\tau_0 e^{-\frac{t}{T}} + A \quad (4)$$

where, $\Delta\tau_0$ represented the magnitude of total stress relaxation and T denoted the viscoelastic time constant, and A was the asymptotic term, representing the stress at the equilibrium stage.

3. RESULTS

The progressive loading scheme effectively revealed progressive changes in the pre- and post-yield behavior of human cortical bone in shear (Fig. 2a). Bone specimens failed along the center line in the gage region connecting the two notches, indicating that the failure occurred in the region where the maximum shear was applied (Fig. 2b). The cortical bone specimens of human tibiae exhibited shear strength of 61.4 ± 6.30 MPa and shear modulus of 4.66 ± 1.10 GPa (Table 1). The shear yield strain and yield stress of the bone specimens were $0.88 \pm 0.18\%$ and 35.7 ± 9.88 MPa, respectively (Table 1).

The progressive change in the shear modulus (G_i) of human cortical bone showed a trend of decay ($G_i = 4.539e^{-19.6\gamma_i}$, $p < 0.001$, Fig. 3) with increasing applied shear strain (γ_i). At 1% of applied strain (γ_i), the shear modulus (G_i) remained about 82% of the initial shear modulus. At 3% of applied shear strain, the shear modulus decreased to about 56% of the initial shear modulus.

A quasi-linear relationship ($\gamma_p = 0.525\gamma_i - 0.521$, $p < 0.001$) was observed between the plastic (γ_p) and applied shear strain (γ_i) (Fig. 4). The plastic deformation was almost indiscernible in the first loading cycle. Starting from the second cycle, appreciable permanent deformations were detected.

The magnitude of stress relaxation ($\Delta\tau_0$) of human cortical bone in shear demonstrated a linear change with increasing applied strain (Fig. 5), whereas the relaxation time constant (T) exhibited a decrease initially and tended to reach a relative plateau afterwards (Fig. 6).

4. DISCUSSION

In this study, flat and smooth failure surfaces were observed along the shear direction in the gage region of the bone specimens (Fig. 2b), suggesting that a uniform shear stress field over the gage region was achieved during the inclined double notch shear tests [15]. In a pure shear mode, progressive changes in the mechanical behavior of bone were investigated in following aspects.

First, the observed mechanical properties of human cortical bone in shear are comparable to those reported in the literature. For instance, the shear strength obtained in this study was 61.4 ± 6.30 MPa, whereas the shear strength obtained from the previous studies ranged from 50.4 ± 14.1 MPa for double-notched shear tests [8], 51.6 ± 1.9 MPa for Acran test and Iosipescu tests [11], 67.6 MPa (41.5 – 105.5 MPa) for double shearing strength tests [17], to 68.0 ± 4.0 MPa [18] and 74.1 ± 3.2 MPa [9] for torsional tests. Furthermore, the shear modulus observed in this study is 4.66 ± 1.10 GPa, which is also in the close range of values reported in literature. For example, the shear modulus of human femoral cortical bone obtained in torsional tests was reported as 3.28 GPa [18], 4.74 ± 0.65 GPa [19], and 5.0 ± 0.2 GPa [9], respectively. In addition, the shear yield strain was observed as $0.88 \pm 0.18\%$ in this study, which is slightly less than ($1.3 \pm 0.1\%$) of human cortical bone specimens obtained in torsional tests [9, 10].

Next, the load-induced modulus loss in human cortical bone was also observed in pure shear tests, which is similar to those observed in tension and compression tests. However, the rate of modulus degradation was much slower in shear mode than that in tension and compression modes. The shear modulus of bone sustains about 82% of initial modulus at 1% of applied strain, whereas the elastic modulus of bone in tension [4] and compression tests [6] drops almost in half (50%) of the initial elastic modulus at the same strain level. At 3% of applied strain, the shear modulus degrades to about 56% of the initial shear modulus in this study, but the elastic modulus under tension [4] and compression modes [6] falls further to about 15% of the initial values. In addition, the damage sensitivity constant (m) of bone is much smaller ($m = 19.6$) in shear than in tension and compression ($m = 64.3$) for age-matched human cortical bone specimens [6, 7]. The above results imply that it is more difficult for microdamage to accumulate in bone specimens under shear load than under tension and compression load. Such conjecture is supported by the observation that resistance to crack growth in human cortical bone is greater in shear than in tension [20]. In addition, microdamage induced by shear is mainly in the form of delamination, which may consequently delay microcrack coalescence and postpone the formation of fatal cracks [10]. Moreover, debonding of interfaces between osteons and the surrounding bone matrix and between osteonal lamellae are always associated with the longitudinal shear failure [8]. Combining the above results, it is presumable that human cortical bone may tend to adapt to prevent crack growth in shear [20].

Furthermore, the yield strain in shear observed in this study was about $0.88 \pm 0.18\%$ ($N = 6$), which is higher than those in compression ($0.71 \pm 0.07\%$, $N = 8$) and in tension ($0.39 \pm 0.03\%$, $N = 8$) when the age-matched bone specimens are compared using the results reported in our previous studies [6, 7]. A student-t test indicated that yield strain in shear is significantly ($p = 0.03$) higher than that in compression. These results indicate that human cortical bone in shear yields at a greater strain than in tension and compression, thus again supporting the aforementioned conjecture that human cortical bone tends to prohibit shear failures.

It is also noteworthy that the quasi-linear relationship between the plastic and applied strain in shear (Fig. 4) is similar with those observed in tension and compression tests [6, 7].

However, the slope of the linear relationship (0.53) is in between the values obtained in tension (0.45) and compression (0.61) tests [6, 7]. Considering the cross-hatch damages in human cortical bone associated with the compression mode [21], it is presumable that shear failure is the major mode in compression tests. This conjecture can be verified by the following examination. First, it is known that the angle of the cross-hatch damages is reported in about 30° with respect to the loading direction [21]. Assuming that plastic deformation in compression is due to the cross-hatch damages and there exists a linear relationship between the plastic with applied strains, the slope of the shear plastic strain vs. applied strain curve would be 0.866 times of the slope of the normal plastic strain vs. applied strain curve. Since the slope of the normal plastic strain curve is 0.61, the estimated slope for the shear plastic strain would be 0.53, which is consistent with the average of measured values obtained in this study. On the other hand, diffuse damage is formed in bone in tension tests, the shear failure might not be the dominant mechanism of post-yield deformation of bone in tension. It is most probably the reason why a much smaller value was observed for the slope of the plastic to applied strain curve in tension mode.

Finally, the viscous behavior (*i.e.* relaxation time constant and magnitude of stress relaxation) of human cortical bone in shear observed in this study (Figs. 5 & 6) differs slightly from the behavior reported in tension and compression modes [6, 7]. The results of this study indicated that the degree of stress relaxation in shear increased continuously with rising applied strain without reaching a plateau in the entire loading process. However, previous studies report that the magnitude of stress relaxation in tension and compression modes increases rapidly within the initial 1% of applied strain and then reaches a plateau after the 1% strain level, showing a saturation of the effect [6, 7]. These results suggest that the viscous behavior of cortical bone is enhanced continuously with increasing strain in shear mode, but not in tension and compression modes. We speculate that shear may cause a progressive deterioration in the collagen phase, thus leading to a continuous increase in the viscous response of bone.

There are several limitations of this study, which include (1) a limited representation of bone sample populations (*e.g.* age, sex, and anatomic locations); (2) the shear tests were conducted only in the longitudinal axis of bone; and (3) the type and form of shear microdamage were not investigated.

CONCLUSIONS

The inclined double notch shear test has been performed to assess progressive changes in the post-yield behavior to human cortical bone in shear using an established progressive loading scheme. The plastic deformation of human cortical bone in pure shear exhibits a quasi-linear relationship with the applied strain, which is similar with those observed in tension and compression tests. However, modulus degradation and viscous behavior of human cortical bone in pure shear mode demonstrate slightly distinct trends compared with those in tension and compression tests. It is speculated that the underlying cause of yielding in human cortical bone is independent of loading modes.

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Fig. 1. Inclined double-notch shear test of human cortical bone: (a) Specimen preparation; (b) Attachment of strain gages to bone specimens; Units are in millimeters. (c) Setup of loading fixture in which the tilting angle (α) is 33.5° . Schematics are not drawn to scale.

(a)

(b)

Fig. 2.

(a) A typical stress-strain curve of progressive loading scheme; (b) Specimen failure resulting from pure shear in the gage region.

Fig. 3. Relationship between the shear modulus (G_i) and applied strain (γ_i) of human cortical bone. Different specimens were represented by various marker styles.

Fig. 4. Relationship between plastic (γ_p) and applied shear strain (γ_i) of human cortical bone. Different specimens were represented by various marker styles.

Fig. 5. Magnitude of stress relaxation ($\Delta\tau_0$) as a function of applied strain (γ_i). Different specimens were represented by various marker styles.

Fig. 6. Relationship between the viscoelastic time constant (T) and applied strain (γ_i) of human cortical bone. Different specimens were represented by various marker styles.

Table 1Descriptive statistics for shear properties of human cortical bone under progressive loading (mean $\pm$ s.d.)

G_θ (GPa)	τ_y (MPa)	τ_u (MPa)	γ_y (%)
4.66 $\pm$ 1.10	35.7 $\pm$ 9.88	61.4 $\pm$ 6.30	0.88 $\pm$ 0.18