Journal of Biomechanics 40 (2007) 2898–2903

JOURNAL OF BIOMECHANICS

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On shear properties of trabecular bone under torsional loading: Effects of bone marrow and strain rate

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Accepted 4 March 2007

Abstract

To further improve our understanding of trabecular bone mechanical behavior in torsion, our objective was to determine the effects of strain rate, apparent density, and presence of bone marrow on trabecular bone shear material properties. Torsion tests of cylindrical trabecular bone specimens from sheep lumbar vertebrae with and without bone marrow were conducted. The bones with marrow were divided into two groups and tested at shear strain rates of 0.002 and 0.05 s^{-1} measured at the specimen perimeter. The bones without marrow were divided into three groups and tested at shear strain rates of 0.002, 0.015, and 0.05 s^{-1}. Comparing the results of bones with and without marrow tested at low (0.002 s^{-1}) and high (0.05 s^{-1}) strain rates, presence of bone marrow did not have any significant effect on trabecular bone shear modulus and strength. In specimens without marrow, power relationships were used to define shear strength and modulus as dependent variables in terms of strain rate and apparent density as independent variables. The shear strength was proportional to the apparent density raised to the 1.02 power and to the strain rate raised to the 0.13 power. The shear modulus was proportional to the apparent density raised to the 1.08 power and to the strain rate raised to the 0.07 power. This study provides further insight into the mechanism of bone failure in trauma as well as failure at the interface between bone and implants as it relates to prediction of trabecular bone shear properties.

Keywords: Trabecular bone; Torsion; Strain rate; Bone marrow; Shear properties

1. Introduction

Trabecular bone is a porous structure consisting of a bony network of connecting rods or plates (solid phase) filled with bone marrow (fluid phase). Trabecular bone exhibits viscoelastic behavior, which depends on the material behavior of its bone tissue as well as the interaction between its bone and bone marrow (solid and fluid phases).

The viscoelastic behavior of bone tissue has been demonstrated by the influence of strain rate on its strength and stiffness (Carter and Hayes, 1977; Linde et al., 1991). Carter and Hayes (1977) performed compression tests on trabecular bone samples at different strain rates and analysed a power relationship of stiffness and strength vs. strain rate and apparent density. They found that both stiffness and strength were proportional to strain rate to the 0.06 power. Linde et al. (1991) discussed different curve fitting methods in defining different power relationships with different powers for strain rate and apparent density. Using a multiple variable regression technique, Linde et al. (1991) reported the following power relations for compressive strength (σ_u) and elastic modulus (E) as dependent variables and apparent density (ρ) and strain rate ($\dot{\varepsilon}$) as independent variables:

$$\sigma_u = 40.2 \times \rho^{1.65} \times \dot{\varepsilon}^{0.073}, \quad (1)$$

$$E = 2232 \times \rho^{1.56} \times \dot{\varepsilon}^{0.047}. \quad (2)$$

Hydraulic stiffening of trabecular bone due to the presence of bone marrow has also been investigated employing both finite element (Kasra and Grynpas, 1998;

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0021-9290/$ - see front matter © 2007 Published by Elsevier Ltd.
doi:10.1016/j.jbiomech.2007.03.008

Simon et al., 1985) and experimental methods (Ochoa et al., 1991; Carter and Hayes, 1977). Ochoa et al. (1991) measured femoral head stiffness and studied its changes by altering the intraosseous fluid boundary condition. They suggested that a mechanical strengthening mechanism due to fluid may be present in intact trabecular bone, and the overall stiffness reflects the material properties of both the porous solid matrix and the entrapped fluid. Carter and Hayes (1977) showed that at very high strain rates ($10.0\,s^{-1}$) and in a confined testing condition the effect of bone marrow in increasing bone strength and stiffness becomes significant. The previous studies mainly investigated the behavior of trabecular bone under compressive loading. The behavior of trabecular bone under other in vivo loading conditions such as torsion inducing shear stress and strain within the bone tissue still needs further investigation.

The mechanical behavior of trabecular bone under torsional loading may play a critical role in the etiology of age-related bone fractures and prosthesis loosening. In the proximal femur, loading associated with falls can induce high shear stresses in trabecular bone (Lotz et al., 1991). Sammarco et al. (1971) reported some shift in the position of the torsional fracture in long bones from the center of the bone to the ends as loading rates increased. This change in location was suggested to be due to the change in shear strength of trabecular bone as compared to cortical bone at different loading rates. The shear strength of trabecular bone also affects implant stability, since mechanical loosening of the implant is likely to occur through the failure in shear at the bone–implant interface (Huiskes, 1990). In this case, shear strength of trabecular bone must be regarded as of more importance than its compressive strength.

Previous investigations of the mechanical properties of trabecular bone have included a few determinations of its shear strength and modulus by conducting pure shear (Stone et al., 1983; Halawa et al., 1978; Saha and Gorman, 1981; Mitton et al., 1997) and torsion tests (Wang et al., 2005; Bruye're et al., 1999; Ford and Keaveny, 1996; Ashman et al., 1987). Considering trabecular bone shear properties there is no power equations yet defined having shear strain rate as an independent variable, and the reported relations only relate shear strength to apparent density. Stone et al. (1983) tested bovine trabecular bone specimens from the proximal humerus in pure shear. They found shear strength to be proportional to apparent density to the exponent 1.65. Testing bovine femoral cancellous bones in torsion, Ashman et al. (1987) found shear modulus to be proportional to structural density to the exponent 1.209. Saha and Gorman (1981) tested trabecular bone samples from human femoral head in pure shear at different loading rate. They found the shear strength to be a function of the strain rate; however, they did not define any power relation similar to those defined for compressive strength.

The objective of this investigation was to address the influence of strain rate and bone marrow in trabecular bone shear properties as well as description of its shear strength and shear modulus as a function of apparent density and the applied loading rate. Power relations can be used to predict trabecular bone shear properties similar to the relations defined to predict its compressive mechanical properties (Keller, 1994).

2. Materials and methods

Thirteen fresh L2–L3 lumbar segments from 11 young (6 months old) and two mature (2 years old) female merino sheep were harvested for this study and stored at −40 °C for 2 months before the processing. Fifty-two cylindrical specimens, from left and right lateral regions of L2 and L3 vertebrae of each sheep (four from each sheep) were cut using a diamond core drill under continuous irrigation (Fig. 1). The specimens were cored in the cranial–caudal direction with their main trabecular orientation aligned in the longitudinal axis of the core and were approximately transversely isotropic (Mitton et al., 1997; Augat et al., 1998). Each specimen had a diameter of 5 mm and a length of 20 mm. Using a water jet, bone marrow was removed from all the specimens except 16 of those belonging to the young sheep. All specimens were processed in the same way and kept hydrated during preparation and testing procedure using saline solution. After preparation, specimens were stored at −20 °C in air-tight vials for a week until mechanical testing. Before testing, apparent densities (ρ) of the bones without marrow were measured by using an analytical balance as described in the literature (Kasra and Grynpas, 1995). Using set screws, the ends of each specimen were fixed in the rectangular cavities of two brass shafts installed in a custom-made torsion

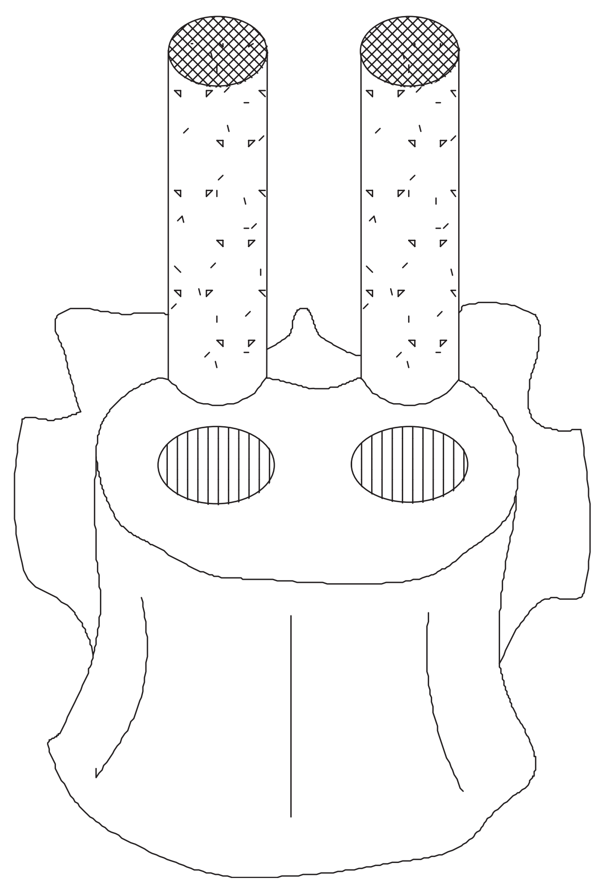

Fig. 1. Vertebral specimens were cored in the cranial–caudal direction from right and left lateral regions of each vertebral body. The specimens were transversely isotropic and the main trabecular orientation was aligned with the longitudinal axis of the core.

machine, and the cavities were filled with PMMA (Fig. 2). In this case, all the specimens would fracture within the specimen gauge length and not at the fixation ends. The bones were then tested to failure in torsion at three deformation rates of 0.013, 0.065, and 0.365 rad s^{-1}. From the slope of the linear region of the recorded torque–angular deformation curve (Fig. 3), the torsional stiffness, K, was measured.

$$K = \frac{\Delta T}{\Delta \theta}. \tag{3}$$

Shear modulus (G) was then calculated as

$$G = \frac{K \times L}{J}, \tag{4}$$

where T is the applied torque, θ is the angular deformation, L is the gauge length, and J is the polar moment of inertia. Eqs. (3) and (4) or their equivalents have been used in calculations of torsional stiffness and shear modulus of cylindrical trabecular bone specimens of ~5–8 mm diameter in different studies (Ashman et al., 1987; Ford and Keaveny, 1996; Bruye're et al., 1999; Wang et al., 2005). In these studies, the specimens could be considered as a continuum solid bar with transverse isotropy allowing the use of Eqs. (3) and (4) with minimal error (Ashman et al., 1987). Trabecular bone structure can be considered a continuum if it has more than five intratrabecular spaces along its dimensions (Harrigan et al., 1988; Kasra and Grynpas, 1998). Having at least ~7 intratrabecular spaces in a diameter of 5 mm, our specimens could be considered a continuum.

For an isotropic material shear stress (τ) can be determined from the torque–angular deformation curve using the following equation (Nadai, 1950).

$$\tau = \frac{1}{2\pi R^3}\left[\phi \frac{\mathrm{d}T}{\mathrm{d}\phi} + 3T\right], \tag{5}$$

where ϕ is the angular deformation per unit length ($\phi = (\theta/L)$) and R is the specimen radius. This equation has been used in determining shear stress of approximately transversely isotropic trabecular bone cylindrical samples (Ford and Keaveny, 1996; Bruye're et al., 1999; Wang et al., 2005). Using Eq. (5), maximum shear stress (τ_{max}) representing shear strength was calculated by substituting the torque value (T) with its recorded maximum value (T_{max}) at the peak point of the torque–angular deformation curve (Fig. 3) where the slope was zero $((\mathrm{d}T/\mathrm{d}\phi) = (\mathrm{d}T/\mathrm{d}\theta) = 0)$, i.e.

$$\tau_{max} = \frac{3T_{max}}{2\pi R^3}. \tag{6}$$

Shear strain rate ($\dot{\gamma}$) was calculated as

$$\dot{\gamma} = \frac{R \times \dot{\theta}}{L}, \tag{7}$$

where $\dot{\theta}$ is the deformation rate. The shear strain rate and maximum shear stress calculated in the above equations correspond to the points on the surface of the cylindrical specimen located at a distance R from the torsion axis.

The bones with marrow (WM) were divided into two groups and tested at strain rates ($\dot{\gamma}$) of 0.002 s^{-1} (Group 1, $n = 8$) and 0.05 s^{-1} (Group 2, $n = 8$). The bones without bone marrow (NM) were divided into three groups and tested at strain rates of 0.002 s^{-1} (Group 3, $n = 12$), 0.015 s^{-1} (Group 4, $n = 12$), and 0.05 s^{-1} (Group 5, $n = 12$). The allocation of specimens into the groups was done in such a manner that the specimens in each group included one specimen from each sheep. Using the results of the specimens without bone marrow (Groups 3–5), a multiple variable regression technique was used to find power relations describing the maximum shear stress (τ_{max}) and shear modulus (G) as a function of apparent density (ρ) and strain rate ($\dot{\gamma}$).

In studying the effect of bone marrow, only specimens belonging to the young sheep were used. In order to investigate whether by increasing the loading rate the presence of bone marrow becomes significant, we compared shear properties of the bones with and without marrow at the slow (0.002 s^{-1}) and high (0.05 s^{-1}) strain rates, i.e. Group 1 ($n = 8$, WM) vs. Group 3 ($n = 8$, NM) tested at the slow, and Group 2 ($n = 8$, WM) vs. Group 5 ($n = 8$, NM) tested at the high strain rate. Each two groups tested at the same strain rate had specimens from the same vertebrae with almost similar densities. In this case, any change in the bone properties was most likely due to the effect of bone marrow and not the density.

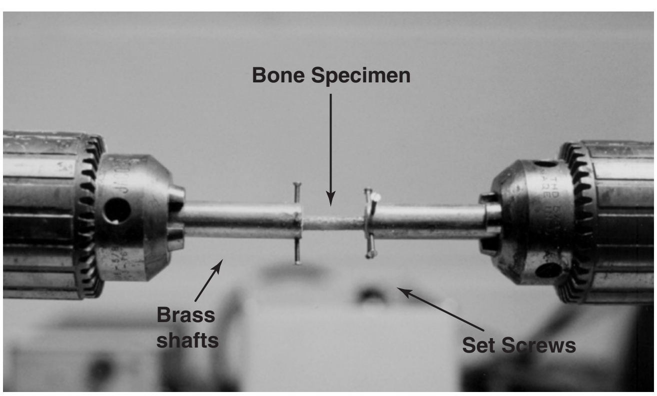

Fig. 2. Trabecular bone fixation assembly for torsion test. The ends of specimens were placed in the cavities at the tips of the brass shafts. They were then fixed by set screws and reinforced using PMMA.

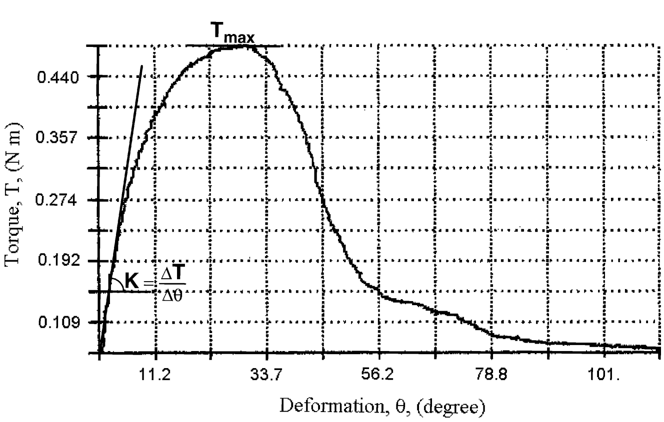

Fig. 3. A typical torque–angular deformation curve. The maximum torque (T_{max}) was the peak value and the torsion stiffness (K) was the slope of the linear region of the curve.

3. Results

As shown in Table 1, the results show no significant differences in shear modulus (G) and shear strength (τ_{max}) between the bones with and without bone marrow tested at the same, low (rows 2 and 3) or high (rows 4 and 5), strain rates. Therefore, presence of bone marrow did not cause any stiffening effect (increase in shear modulus) by increasing the strain rate. However, comparing the results of different strain rates (rows 2 and 3 vs. rows 4 and 5), there were significant increases in shear modulus and strength by increasing the strain rate showing the viscoelastic effect of the bone tissue.

Fig. 4 shows the variation of shear strength against apparent density on log–log scale for different strain rates. This figure represents the power relation between shear strength (τ_{max}) as dependent variable and apparent density

Table 1
Means and standard deviations (mean ± SD) for the mechanical properties of bones with and without marrow at different strain rates

Groups	Apparent density, ρ ($g\,cm^{-3}$)	Strain rate, $\dot{\gamma}$ (s^{-1})	Shear modulus, G (MPa)	Shear strength, τ_{max} (MPa)
With marrow Group 1 ($n = 8$)	–	0.002	$263 \pm 67^{+}$	$4.8 \pm 0.9^{+}$
Without marrow Group 3 ($n = 8$)	0.69 ± 0.07	0.002	$282 \pm 81^{+}$	$4.9 \pm 1.5^{+}$
With marrow Group 2 ($n = 8$)	–	0.05	314 ± 74	$7.3 \pm 1.9^{*}$
Without marrow Group 5 ($n = 8$)	0.77 ± 0.14	0.05	$366 \pm 76^{*}$	$7.7 \pm 1.4^{*}$

In each column, the values indicated with the symbol "*" are significantly higher than those with the symbol "+" ($p < 0.05$).

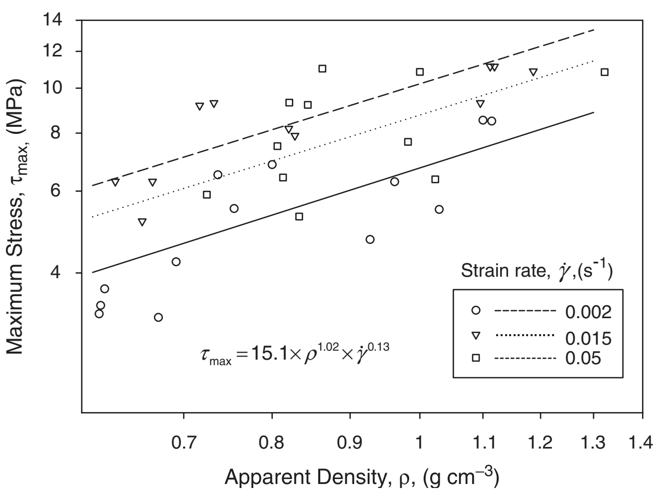

Fig. 4. Variation of shear strength (τ_{max}) against apparent density (ρ) on log–log scale for different strain rates ($\dot{\gamma}$). Multiple regression analysis showed significant dependence of shear strength on both apparent density and strain rate ($p < 0.05$).

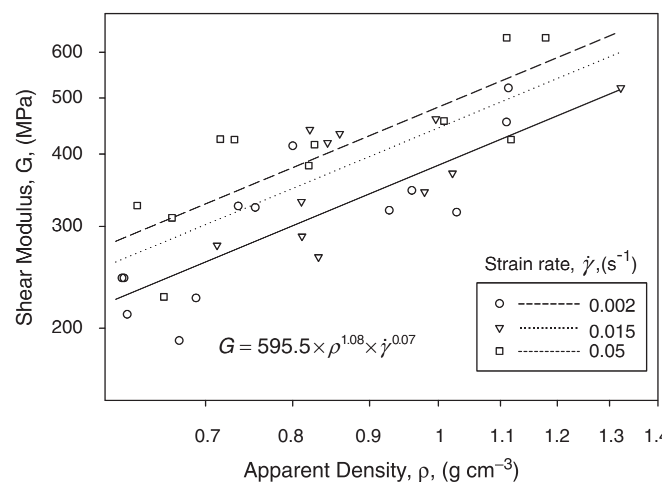

Fig. 5. Variation of shear modulus (G) against apparent density (ρ) on log–log scale for different strain rates ($\dot{\gamma}$). Multiple regression analysis showed significant dependence of shear modulus on both apparent density and strain rate ($p < 0.05$).

(ρ) and strain rate ($\dot{\gamma}$) as independent variables.

$$\tau_{max} = 15.1 \times \rho^{1.02} \times \dot{\gamma}^{0.13} \quad (r = 0.89,\ p < 0.0001). \tag{8}$$

Fig. 5 shows the variation of shear modulus against apparent density on log–log scale for different strain rates. The power relation established between shear modulus (G) as dependent variable and apparent density (ρ) and strain rate ($\dot{\gamma}$) as independent variables was

$$G = 595.5 \times \rho^{1.08} \times \dot{\gamma}^{0.07} \quad (r = 0.84,\ p < 0.0001). \tag{9}$$

Table 2 shows estimated coefficients in regression analysis for Eqs. (8) and (9). Considering a significance level of $p < 0.05$, all the p values associated with the regression coefficients were small enough to conclude that there were no nonsignificant terms in the power relations (8) and (9).

Table 2
Estimated coefficients in regression analysis

Equation no.	Dependent variable (Y)	Estimated values of coefficients[a] (standard error in parentheses)		
		a	b	c
8	τ_{max}	15.1 (1.14), $p < 0.0001$	1.02 (0.19), $p < 0.0001$	0.13 (0.03), $p = 0.0004$
9	G	595.5 (1.11), $p < 0.0001$	1.08 (0.14), $p < 0.0001$	0.07 (0.02), $p = 0.0034$

Equation no. refers to its number in the text.
[a]Coefficients a, b, and c refer to their corresponding values in Eqs. (8) and (9) according to the power relation $Y = a \times \rho^{b} \times \dot{\gamma}^{c}$.

4. Discussion

We performed torsion tests on trabecular bone samples from sheep lumbar vertebrae in order to investigate the effects of bone marrow and strain rate in affecting trabecular bone shear strength and modulus. The values of mean shear strength reported in the literature vary between 3.65 and 7.5 MPa according to different specimen conditions, such as species and anatomical site with mean apparent densities ranging from 0.48 to 0.6 $g\,cm^{-3}$, as well as the testing method (Stone et al., 1983; Halawa et al., 1978; Saha and Gorman, 1981; Mitton et al., 1997; Ford and Keaveny, 1996; Bruye're et al., 1999). In this case, the reported range of mean shear modulus was from 183 to 349 MPa (Wang et al., 2005; Ford and Keaveny, 1996; Ashman et al., 1987). Our mean values of shear strength and modulus (Table 1) compare well with the reported studies.

Within the range of loading rate reported in this study, presence of bone marrow did not affect trabecular bone shear strength and modulus. Experimental (Carter and Hayes, 1977) and finite element studies (Kasra and Grynpas, 1998; Simon et al., 1985) have shown that the hydraulic stiffening of bone marrow occurs when the applied loading rate is greater that the pore fluid diffusion rate (so-called high loading rates). Experimentally, this has only been shown at a very high loading rate and in a confined testing condition under compressive loading (Carter and Hayes, 1977). The entrapped marrow resists compressive forces while it does not seem to resist shear forces substantially. In compression, the change in bone volume with an entrapped bone marrow causes a substantial stiffening effect while in torsion there is relatively no change in bone volume and any stiffening effect is caused by much smaller frictional forces between bone and marrow. In our study, the specimens were only constraint at the ends (Fig. 2) and the bone marrow could flow freely through the open boundary along the length. However, during testing, we observed that the marrow flowed only through a small region of the open boundary at the fracture site. This suggests that confining the bone marrow in torsion likely has less stiffening effect than that in compression where bone marrow can flow through all regions of the open boundary (Carter and Hayes, 1977). Therefore, it is expected that hydraulic stiffening of bone marrow to be less effective in torsion than compression. In this case, a much higher loading rate than those of this study may be required to show any stiffening response due to the presence of bone marrow.

Under compressive loading, the power of the apparent density and strain rate in the relationships defining compressive strength (σ_u) and elastic modulus (E) as dependent variables and apparent density (ρ) and strain rate ($\dot{\varepsilon}$) as independent varies, depend on the curve fitting method used (Keaveny and Hayes, 1993; Carter and Hayes, 1977; Linde et al., 1991) and also the type of cancellous bone, plate-like or rod-like (Gibson, 1985). For example, Carter and Hayes (1977) defined a cubic relation between modulus and apparent density, while Keaveny and Hayes (1993) reported an exponent of 1.44 for the apparent density. This is also expected to be the case when defining power equations relating bone shear properties in terms of density and strain rate.

Trabecular bone power–law relationship between shear properties and apparent density can be anticipated from the foam analogy. Experimental studies of the shear strength of foams have shown the exponent of apparent density to have an average value of 1.1 (Harding et al., 1965; Traegar, 1967). In this case, the average value of the exponent of apparent density for shear modulus was 1.08 (Harding et al., 1965). In the same studies of foam structures, the average values of the exponents of apparent density for compressive strength and modulus were both 1.6 (Traegar, 1967) or 1.4 (Harding et al., 1965), which were higher than those of shear strength and modulus. Patel (1969) also theorized that the shear strength of open-celled foams would be related to apparent density by a power law with an exponent of approximately 1.0. Therefore, the exponents of apparent density for defining trabecular bone shear strength and modulus found in this study (Eqs. (8) and (9)) as well as those reported by Linde et al. (1991) for compressive properties (Eqs. (1) and (2)) are reasonable in light of the values reported for foams (Harding et al., 1965; Traegar, 1967).

We used a multiple variable regression technique in defining the power relations for trabecular bone shear properties (Eqs. (8) and (9)) similar to that of Linde et al. (1991) used to define compressive properties (Eqs. (1) and (2)). Comparing the relations of compressive loading (Linde et al., 1991) with those found in this study for torsional loading (Eqs. (1) vs. (8) and (2) vs. (9)), the powers of density were larger in compression that those in torsion while the powers of strain rate was larger in torsion than those in compression. For example, comparing Eqs. (1) vs. (8), the power of density (ρ) for compressive strength (σ_{max}) was 62% larger than that of shear strength (τ_{max}) while the power of shear strain rate ($\dot{\gamma}$) was 78% larger than that of compressive strain ($\dot{\varepsilon}$). Therefore, compared with compressive loading, it is likely that the effect of density is weaker and the effect of strain rate is stronger in changing trabecular bone shear properties. In this case, we did not perform any compression tests and compared our torsion data with compression data reported by Linde et al. (1991) using different type of bone samples. Therefore, this argument may be further verified using the same batch of bone samples for both torsion and compression tests. With this consideration, our results suggest that trabecular bone shear strength and modulus are likely less dependent on apparent density and more dependent on strain rate compared to its compressive strength. Therefore, its shear properties are likely less vulnerable to bone loss than its compressive properties.

Acknowledgments

We would like to thank Dr. George Tomlinson for his assistance in statistical analyses. This work was supported by Canadian Institute of Health Research (CIHR).

References

Ashman, R.B., Corin, J.D., Turner, C.H., 1987. Elastic properties of cancellous bone: measurement by an ultrasonic technique. Journal of Biomechanics 20, 979–986.

Augat, P., Link, T., Lang, T.F., Lin, J.C., Majumdar, S., Genant, H.K., 1998. Anisotropy of the elastic modulus of trabecular bone specimens from different anatomical locations. Medical Engineering & Physics 20, 124–131.

Bruye're, K., Dumas, R., Rumelhart, C., Arlot, M.E., 1999. Mechanical characterization in shear of human femoral cancellous bone: torsion and shear tests. Medical Engineering & Physics 21, 641–649.

Carter, D.R., Hayes, W.C., 1977. The compressive behaviour of bone as a two-phase porous structure. Journal of Bone and Joint Surgery 59A, 954–962.

Ford, C.M., Keaveny, T.M., 1996. The dependence of shear failure properties of trabecular bone on apparent density and trabecular orientation. Journal of Biomechanics 29, 1309–1317.

Gibson, L.J., 1985. The mechanical behaviour of cancellous bone. Journal of Biomechanics 18, 317–328.

Halawa, M., Lee, A.J.C., Ling, R.S., Vangala, S.S., 1978. The shear strength of trabecular bone from the femur, and some factors affecting the shear strength of the cement–bone interface. Archives of Orthopaedics and Trauma Surgery 92, 19–30.

Harding, R.H., Hastettler, F., Mahoney, T.J., 1965. Handbook of Foamed Plastics. Lake publishing, Libertyville, IL.

Harrigan, T.P., Jasty, M., Mann, R.W., Harris, W.H., 1988. Limitations of the continuum assumption in cancellous bone. Journal of Biomechanics 21, 269–275.

Huiskes, R., 1990. The various stress patterns of press-fit, ingrown, and cemented femoral stems. Clinical Orthopaedics 261, 27–38.

Kasra, M., Grynpas, M.D., 1995. The effects of androgens on the mechanical properties of primate bone. Bone 17, 265–270.

Kasra, M., Grynpas, M.D., 1998. Static and dynamic finite element analyses of an idealized structural model of vertebral trabecular bone. Journal of Biomechanical Engineering 120, 267–272.

Keaveny, T.M., Hayes, W.C., 1993. A 20-year perspective on the mechanical properties of trabecular bone. Journal of Biomechanical Engineering 115, 534–542.

Keller, T.S., 1994. Predicting the compressive mechanical behaviour of bone. Journal of Biomechanics 27, 1159–1168.

Linde, F., Nørgaard, P., Hvid, I., Odgaard, A., Søballe, K., 1991. Mechanical properties of trabecular bone. Dependency on strain rate. Journal of Biomechanics 24, 803–809.

Lotz, J.C., Cheal, E.J., Hayes, W.C., 1991. Fracture prediction for the proximal femur using finite element models: Part I—Linear analysis. Journal of Biomechanical Engineering 113, 353–360.

Mitton, D., Rumelhart, C., Hans, D., Meunier, P.J., 1997. The effects of density and test conditions on measured compression and shear strength of cancellous bone from the lumbar vertebrae of ewes. Medical Engineering & Physics 19, 464–474.

Nadai, A., 1950. Torsion of a Round Bar. The Stress–Strain Curve in Shear. In Theory of Flow and Fracture of Solids. McGraw-Hill, New York.

Ochoa, J.A., Heck, D.A., Brandt, K.D., Hillberry, B.M., 1991. The effect of intertrabecular fluid on femoral head mechanics. The Journal of Rheumatology 18, 580–584.

Patel, M.R., 1969. The deformation and fracture of rigid cellular plastics under multiaxial stress. Ph.D. Thesis, University of California, Berkeley.

Saha, S., Gorman, P.H., 1981. Strength of human cancellous bone in shear and its relationship to bone mineral content. Transactions of the 27th Annual Orthopaedic Research Society, p. 217.

Sammarco, G.J., Burstein, A.H., Davis, W.L., Frankel, V.H., 1971. The biomechanics of torsional fractures: the effect of loading on ultimate properties. Journal of Biomechanics 4, 113–117.

Simon, B.R., Wu, J.S.S., Carlton, M.W., Evans, J.H., Kazarian, L.E., 1985. Poroelastic dynamic structural models of rhesus motion segments. Spine 10, 494–507.

Stone, J.L., Beaupre, G.S., Hayes, W.C., 1983. Multiaxial strength characteristics of trabecular bone. Journal of Biomechanics 16, 743–752.

Traegar, R.K., 1967. Physical properties of rigid polyurethane foams. Journal of Cellular Plastics 3, 405–418.

Wang, X., Guyette, J., Liu, X., Roeder, R.K., Niebur, G.L., 2005. Axial–shear interaction effects on microdamage in bovine tibial trabecular bone. European Journal of Morphology 42, 61–70.