Research Paper

Effect of torsional loading on compressive fatigue behaviour of trabecular bone

S.J. Fatihhi\textsuperscript{a}, A.A.R. Rabiatiul\textsuperscript{a}, M.N. Harun\textsuperscript{a,b}, Mohammed Rafiq Abdul Kadir\textsuperscript{c}, T. Kamaruld\textsuperscript{d}, Ardiyansyah Syahrom\textsuperscript{a,b,*}

\textsuperscript{a}Faculty of Mechanical Engineering, Universiti Teknologi Malaysia, Johor, Malaysia
\textsuperscript{b}Sport Innovation and Technology Centre (SITC), Institute Human Centred Engineering (IHCE), Universiti Teknologi Malaysia, Johor, Malaysia
\textsuperscript{c}Medical Devices and Technology Group (Mediteg), Faculty of Biomedical Engineering and Health Science, Universiti Teknologi Malaysia, Johor, Malaysia
\textsuperscript{d}Tissue Engineering Group, NOCERAL, Department of Orthopaedic Surgery, Faculty of Medicine, University of Malaya, 50603 Kuala Lumpur, Malaysia

ARTICLE INFO

Article history:
Received 15 June 2015
Received in revised form 31 August 2015
Accepted 4 September 2015
Available online 14 September 2015

Keywords:
Trabecular bone
Fatigue
Multiaxial
Effect of torsion
Combined loading

ABSTRACT

The present study reports the effects of combined torsional and compressive cyclic loading on trabecular bone in order to mimic true physiological conditions and thereby provides improved data that represents clinical and real life conditions. However, only compressive behaviour is evaluated in most previous studies of bone mechanics. From the monotonic evaluation, it is observed that lower stress is needed for the onset of microcrack in the sample under torsional loading, compared to the stress needed in compression. Trabecular bone samples were subjected to a combination of torsion and compression fatigue at different stress levels during which they were compared to compressive axial fatigue. The stress levels were determined by considering the monotonic strength at 25–50% for both compressive and shear stresses. Significant decrease in fatigue lifetime is observed in between samples of pure compression fatigue and those with superpositioned torsional loading (p<0.05). The reduction in fatigue lifetime became more evident at a high torsional stress level. In this case, the failure of the sample is said to be ‘torsional dominant’. Fatigue behaviour of bovine trabecular bone begins with plastic deformation, followed by strain accumulation and modulus reduction. As the strain rate increases, more energy dissipates and the sample finally failed. Further, the analysis of fractograph revealed something on the trabeculae by bending in sample with superpositioned torsional loading. In conclusion, torsional loading decreases the quality of the trabecular properties in terms of stiffness, life and structural integrity. It is hoped that results from this study will improve the understanding of the behaviour of trabecular bone under combined fatigue and help to develop future assessments of trabecular failure.

© 2015 Elsevier Ltd. All rights reserved.
1. Introduction

Bone fatigue can occur in physiological conditions, even in the absence of bone-related pathological conditions (Martin, 2003). A relatively short lifespan is observed in bones in vivo and this has been associated with the torsional force during overpronation (Holden and Cavanagh, 1991). Thus, in order to evaluate bone behaviour during fatigue, the in vitro assessment should not be restricted to uniaxial loading only. The present work attempts to quantify the fatigue behaviour of bovine trabecular bone under combined compression–torsion. This study focuses on the effect of applied torsional stress on the compression fatigue properties of the trabeculae. Previous studies have demonstrated increased bone fracture risk with associated fatigue deformation (Lin et al., 1994; Matheson et al., 1987; Pozderac, 2002; Williams et al., 2002). However, only few studies are available in which fatigue behaviour of trabecular bone is investigated (Haddock et al., 2004; Michel et al., 1993).

Michel et al. (1993) conducted a fatigue compression experiment on bovine trabecular bone. Their results have demonstrated the relationship between trabecular fatigue life and the initial maximum strain. Observations were also made on the modulus degradation behaviour and failure modes. High-cycle fatigue and low-cycle fatigue exhibited distinct modulus degradation pattern which were accompanied by non-linear increase and hysteresis as well as secant modulus reduction. Slight increase in modulus is observed at the early stage of fatigue for low stress levels, while high stress levels exhibited gradual decrement of modulus. Subsequently, rapid drops of the modulus were demonstrated for both low and high stress levels. Two failure modes were indicated—the brittle-like failure and the buckling-like failure. The trabecula was found to fail by cyclic compression, even within its elastic limit.

Further Moore and Gibson (2003), evaluated the changes in mechanical properties of bovine trabecular bone before and after fatigue compression at different normalised stress levels. Their acclaimed findings agreed with the strain-based failure criterion for bone under which the developed maximum strain better indicates normalised modulus than the number of loading cycles. Failure of the trabecula was suggested to be strain based, as its mechanical properties depend mainly on maximum compressive strain.

On the other hand, Haddock et al. harvested trabecular bone samples from human cadaveric donors which were then subjected to compression fatigue. The S–N curves obtained were then quantitatively compared with the data from literature, in which the samples are of the other species. They found that the secant modulus reduction and strain accumulation seen in human vertebral bone were in agreement with the reported data for bovine tibial bone. With regard to the monotonic yield strains differences, fatigue behaviour for both types of trabecular bone was found to be similar.

Recently, there has been significant interest in the behaviour of trabecular bone towards multiaxial loading (Niebur et al., 2002; Rincón-Kohli and Zysset, 2009). This is important as the uniaxial evaluation is found to overestimate failure and does not closely simulate in vivo loading conditions (Fatihhi et al., 2015). Understanding the function and mechanical properties of the bone, particularly the trabecular in similar environment as the material during service, is useful in the development of bone analogous materials. To the best of our knowledge, none of the fatigue studies on trabecular bone has ever considered multiaxial loading due to the complexities of experimental procedure. However, previous investigations have provided evidences of the effects of multiaxial loading expulsion on bones (Carter, 1978; Holden and Cavanagh, 1991) with exaggerated impact among athletes and military recruits (Bergmann et al., 2001, 2004, 2010; Burr et al., 1996; Matheson et al., 1987). Thus, understanding the mechanical behaviour of the trabecular bone under multiaxial fatigue is necessary. Furthermore, such information is useful in the design and development of bone analogous materials.

Based on the points discussed above, the torsional loading effects on the trabecular bone fatigue compression properties are investigated accordingly. In this study, cleaned trabecular bone with dimensions as described in Section 2 was employed as the test piece. Fatigue test was performed in compression with combination of torsional loading. Compression fatigue characteristics of the trabecula under combined loading were targeted for investigation. In other words, the effect of torsional stress on compression fatigue properties of the trabecula is particularly of interest and focus in this study. In vitro fatigue compression and compression–torsion are conducted with systematic variation in relative magnitudes of compressive and torsional loading. The fatigue life of the trabecula is believed to reduce with superposition of torsional loading on fatigue compression. It is hoped that the data provided in this study provides evidence for and enhances knowledge of the effects of these forces and the associated failure management, thereby ensuring further understanding of the trabecular biomechanical responses.

2. Materials and methods

2.1. Sample preparation

Trabecular bone used in this study was from bovine proximal and distal femur (Lotz et al., 1995) obtained from a local slaughterhouse. Although bovine trabecular bone has higher strain resistant in comparison to human bone due to higher bone density, the deformation behaviour has been found to be similar (Haddock et al., 2004) with fatigue strength almost identical to that of the human (Taylor, 2000). Thus, bovine bone is considered as an appropriate representation of human bone for biomechanical studies.

A total of 74 femurs from bovine aged two to three years old were dissected using a ±150 rpm diamond saw (Behringer GmbH, typeSLB 230 DG HA, Kirchardt). In the present work, samples were taken only from medial–lateral condyle region. The dissected bone was then drilled into 119 samples of 10 mm diameter using a 1.5 mm thick diamond–tip coring bit at 150–250 rpm (Shim et al., 2005), oriented parallel to the long axis of the bone (Fig. 1(a)). During machining, the bone was kept wet with saline solution and the temperature was monitored not to exceed 40 °C (Fluke 62 Mini Infra-red Thermometer) to prevent...
heat-related damage (Gronkiewicz et al., 2009) whilst the room temperature was maintained at 24 °C. The samples were then cut into the final total length of 35 ± 5 mm. The marrow trapped in the trabecular samples was removed by ultrasonic cleaning in water bath (Cest ultrasonic, model P1100SR, USA) at 37 °C (Shim et al., 2005) with added detergent solution (Pumicized, Gent-l-kleen, USA) at a later stage of the cleaning process (Kohles et al., 2001). The remaining marrow was further cleansed using water and air jets to reduce microarchitectural damage due to freezing fluid expansion during freezing of bone. The removal of marrow is intended to facilitate the micro-structural analyses of the sample before and after mechanical test. All the prepared samples were wrapped in saline dampened gauze and stored in airtight container at −20 °C (Teo et al., 2007; van Lenthe et al., 2006).

Analytical balance with Archimedes’s method was used to determine the porosity and apparent density of the samples. Both extremities of the samples were embedded in blocks of polymer material, following a common procedure (Beaugier et al., 2007; Lind et al., 2001; Rincón-Kohli and Zysset, 2009; Sharir et al., 2008). The polymeric material was a mixture of resin and hardener (Mecaprex MA2) in the proportion of 3:1. Careful mounting was needed to guarantee perfect alignment between the polymer and the sample. The alignment of the samples was secured by a custom rig. The mounted polymer covered approximately 7–10 mm in length at both sample ends, reducing the sample’s effective length to 20 ± 5 mm. All mechanical testing were conducted at room temperature, thus the samples were thawed to let the surface temperature increase for a few hours. Both mounted extremities of the sample were then fitted to a pair of stainless steel end caps and glued using cyanoacrylate adhesive (Loctite 401, Newington, CT, USA). The sample fixation to the end caps was tightened with screw pin at two opposite sides of the end caps. The end caps of the sample were clamped into the hydraulic wedge grips (Fig. 1(b)) with fixed lower grip and rotatable upper grip.

2.2. Mechanical testing

2.2.1. Monotonic testing
Sixteen samples were employed each for monotonic compression and torsion. These tests were conducted using a universal testing machine (The FastTrack 8874, Instron, Norwood, USA). The strain rate was controlled at 0.001 s⁻¹ for compression test (Turner, 1989), and at 0.02 s⁻¹ for torsion test (Bruyere Garnier et al., 1999). The relative displacement between the end caps was measured using an extensometer (Instron 2620 Series Dynamic Extensometer with travel 5 mm/mm and gauge length 12.5 mm/mm). The initial yield stress was determined by 0.2% offset method and the ultimate stress was the maximum stress value each from pure compression and torsion test. Failure in torsional stress was determined by 0.4% offset line (Bruyere Garnier et al., 1999; Ford and Keaveny, 1996). Equations involved in computing shear stress and stiffness in pure torsion tests are presented in Appendix section.

2.2.2. Fatigue testing
A total of 89 samples were tested for compression, torsion and combined compression–torsion fatigue. Tests were conducted in the same condition as per monotonic testing, except for the type of load cell used (Biaxial Dynacell™, Instron, Norwood, USA). This dynamic load cell has load capacity of 10 kN, torque capacity of 100 Nm and 0.5% precision from 1% of the full scale. Combined fatigue tests implemented typical triangular wave-form as illustrated in Fig. 2. With regard to the absence of a biological response, a maximum number of cycles for all fatigue tests was defined to be 1 million cycles or at fracture, whichever was first to occur. Force and displacement data collected during fatigue were at 100 Hz per cycle, and at 1000 Hz every tenth cycle and during the last 10 cycles (8800 digital controller, Console Software, Instron, Norwood, USA). Automatic routine at selective acquisition of load–displacement data was triggered at number of cycle, N, with N incremented by unity (N = 1, 2, ..., 8, 9, 10, 20, ..., 80, 90, 100, 200, ..., 800, 900, 1000, 2000, ...).
Dynamic Testing Software, Instron, Norwood, USA). Under combination test, initial modulus for both compression and torsion were fitted within the average of 10 cycles. Each sample was loaded at a constant loading rate at frequency of 2 Hz, which simulated the normal human gait pattern (Bergmann et al., 2001). The waveform used here was simplified due to the nature of gait waveform which is too complex to be executed experimentally. Furthermore, only compression and torsional loading were chosen to represent the dominant loading modes exhibited during physiological activities.

Twenty-five percent and 50% of maximum ultimate compressive stress (UCS) and maximum shear stress (USS) from monotonic tests were chosen respectively to represent the threshold limit for shear stress and normal stress expected in vivo (Vashishth et al., 2001). From the load–displacement data, initial modulus of each sample was calculated using the method shown in Fig. 3. Fatigue tests were performed in the constant amplitude tests for each sample and were loaded at various stress levels from 25% to 50% stress from both monotonic compression and torsion. The combined fatigue stress levels are tabulated in Table 1. Failure was defined at 10% reduction in secant modulus in both compressive and torsional fatigue loading in which a sample could no longer sustain the applied stress as indicated by a rapid increase in strain on the subsequent loading cycle. Ten percent reduction in modulus is considered the average for both secant modulus reduction and increase in strain rate before final failure.

2.3. Fracture surface and microstructure

Scanning electron microscopy (SEM) with backscattered image (BSI) analyses was conducted to observe the fracture surfaces and microdamage of samples (Zeiss Supra 35VP). Prior to SEM, bone samples were glued to an aluminium plate and coated with a conductivity layer.

2.4. Statistical analysis

All data is presented in mean and standard deviation. One-way ANOVA, student t-test and regression analyses were performed on the data to determine any significant variations between two groups. This statistical analysis was performed using Minitab® Release 14.12.0 statistical software (Minitab, USA).

3. Results

Monotonic tests for both compression and torsion were first conducted to determine the trabecular mechanical properties and stress levels for the subsequent fatigue tests. As tabulated in Table 2, no difference was found in the apparent density of the bovine trabecular samples selected for both compression (p=0.77) and torsion (p=0.86) testing. This is crucial in order to minimise the effect of sample density on the mechanical properties. However, the elastic modulus of the sample tested in compression as well as the initial yield stress and shear yield strain of the sample testing in torsion were found to be significantly different across the sample. The mean values of ultimate stress for the trabecula under monotonic compression and monotonic torsion were 14.22 MPa and 8.95 MPa, respectively.

The stress levels implemented in fatigue tests were derived from the ultimate stresses obtained from both monotonic compression and torsion. There were two stress levels evaluated for in bovine trabecula under each compression and torsion fatigue, and four stress levels in samples under combined fatigue. The resultant mechanical properties are as shown in Table 3. No difference was found in the apparent density across samples for each stress level (p>0.05).

The damage accumulation progress in the sample tested under fatigue is usually represented by the stress–strain loop translation on the strain axis. Fig. 4 shows the relationships between compressive stress and compressive strain for...
bovine trabecular bone samples under compression fatigue and combined fatigue at different stress levels. Fig. 4(a) and (b) shows the first 10 hysteresis loops for samples under compression fatigue at 25C and 50C respectively. According to both of these figures, the compressive strain is observed to increase with the number of cycles, with a more pronounced increment demonstrated in sample at 25C. However, no clear change in the stress–strain slope is observed in between the
two stress levels. The first 10 hysteresis loops for samples under combined fatigue are shown in Fig. 4(c) and (d). In Fig. 4(c) the increment in compressive strain for sample at 25°C25T is less pronounce compared to the sample at 50°C50T. The change in shape of the stress–strain slope can also be observed clearly.

Due to the absence of significant differences in strain at failure in between samples at 25°C and 50°C (Table 2), data of sample at 50°C were mostly considered herein for comparisons with samples under combined fatigue for ease of presentation. The comparisons were made in order to evaluate the effect of superpositioned torsional loading onto compression fatigue behaviour of bovine trabecular bone. Comparison of the accumulated maximum strain over the lifetime cycles on samples under combined fatigue and that of sample under compression fatigue were shown in Fig. 5. Linear compressive strain is observed over the lifetime of sample under compression fatigue, followed by a significant increase towards the failure of the sample. However, gradual increase in compressive strain is observed in both samples with superpositioned torsional loading at 25T and 50T. In other words, the deformation in compression progressed at small stages throughout the life cycles until the final failure. Furthermore, a more prominent deformation is observed at higher superpositioned torsional loading level.

The resultant fatigue properties of the trabecular samples were further analysed in terms of the fatigue life. The normalised stress on the vertical axis is plotted against the number of cycles to failure on the horizontal axis to better demonstrate the effect of torsional loading on compression fatigue behaviour of the trabecula. The resultant curve from the sample under compression fatigue was compared to those of combined fatigue with superpositioned torsional loading of 25T and 50T (Fig. 6). Here, it can be observed that the stress level has a negative linear correlation towards the fatigue life. Power regression on these curves resulted in the straight lines which were all statistically significant ($p<0.05$). From these lines, the effect of torsional loading can be seen clearly. The lines tend to incline more with increased torsion while the fatigue life shortened with increased stress. Separately, statistical analyses were done in between samples across all stress levels. Decrease in fatigue life of bovine trabecular bone was found to be significant ($p<0.05$) in between samples tested under compression fatigue (25C and 50C) and those with superpositioned torsion counterparts. A more prominent decrease was found in samples with higher torsional load for all combined fatigue stress levels ($p<0.05$). The samples were said to be failed by torsion only at the same or less compression levels as the torsional loading. Otherwise, the decrease in fatigue life was found to have made no difference.

From Fig. 7, estimation can be made to plot the normalised shear stress against the normalised compressive stress as shown in Fig. 8. The S–N characteristics for different normalised stress at cycles to failure at $10^4$, $10^5$, and $10^6$ are investigated. Hill’s equation (Hill, 1948) was implemented to generate arcs on this plot, in which the maximum capability of the fatigue strength for each concerned fatigue life is expressed. In the present work, the experimental results are in agreement with Hill’s criterion, which describes a two-dimensional failure surface and can be expressed as:

$$\left(\frac{\sigma}{\sigma_y}\right)^2 + \left(\frac{\tau}{\tau_y}\right)^2 = 1$$

(1)

where

$\sigma_y$ = apparent compressive yield stress

$\tau_y$ = apparent shear yield stress
In fatigue, Hill’s criterion is expressed as the normalised stress-based formulation:

\[
\frac{\sigma_s}{E_0(N_f)} + \frac{\tau_s}{G_0(N_f)} = 1
\]

where

\[
\sigma_s = \text{maximum cyclic compressive stress}
\]

\[
\tau_s = \text{maximum cyclic shear stress}
\]

\[
E_0 \text{ and } G_0 \text{ are initial modulus and modulus of rigidity respectively.}
\]

The ratio of Young’s modulus at N cycle to Young’s modulus at first cycle is equal to the stiffness of the tested sample. Decrease in sample stiffness can be observed throughout the fatigue life as shown in Fig. 8. The decrease indicated the microstructural damage which was induced by fatigue. It can be seen that the decrease in stiffness becomes more pronounced following the increase in stress amplitudes. For sample under compression fatigue, constant stiffness is demonstrated over the most part of the lifetime (Fig. 8). The decrease was more pronounced with increasing torsional stress levels. For sample under tensile torsion, torsional stiffness is presented in Table 4.

Results from the present study were compared to previous published results from the literature. The comparison is presented in Table 4. Further analysis on the fatigue behaviour of the trabecula was done by characterisation of the fracture surfaces. On the sample loaded by compression fatigue, an oblique-like fracture was exhibited, as shown in Fig. 9(a). The crack line is clearly visible across the top left of the sample to its lower right. At higher magnification, the individual strut of the same sample is observed to suffer microcracks propagation along the line of fibrils (Fig. 9(b)).

The fracture surface of the sample tested under pure compression fatigue was then compared to that of the sample with superpositioned torsional load. The fractograph of the sample tested under combined fatigue is presented in Fig. 10. A fracture line that spirals along the axis of the sample can be observed in Fig. 10(a) with sharp points and edges indicated by the small arrows. The structure along the fracture line seems to be overlapped at some points at the edges (not shown). Higher magnification on the sample surface reveals numerous microcracks developed on individual struts. These microcracks propagated along the line of fibrils and branched out to another line of another strut (Fig. 10(b)).

---

**Fig. 8** – Modulus reduction profile (a) at various stress levels. In (b), the modulus reduction of each loading components of samples under combined compression–torsion is compared at 50C25T (solid lines) and 50C50T (dashed lines).

**Table 4** – The comparison of the lifetime curve obtained from the present study with others from the literature for trabecular bone under uniaxial fatigue compression.

<table>
<thead>
<tr>
<th>References</th>
<th>Site-species</th>
<th>N</th>
<th>Lifetime curve, (\sigma/\sigma_0 = a \times N_f^b)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Michel et al., 1993</td>
<td>Distal femur – bovine bone</td>
<td>24</td>
<td>(\sigma/\sigma_0 = 0.0241 \times N_f^{-0.072})</td>
</tr>
<tr>
<td>Moore and Gibson,</td>
<td>Proximal tibia – bovine bone</td>
<td>100</td>
<td>(\sigma/\sigma_0 = 0.0093 \times N_f^{-0.096})</td>
</tr>
<tr>
<td>2003</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Haddock et al., 2004</td>
<td>Vertebral – human bone</td>
<td>52</td>
<td>(\sigma/\sigma_0 = 0.0093 \times N_f^{-0.117})</td>
</tr>
<tr>
<td>Rapillard et al., 2006</td>
<td>Vertebral – human bone</td>
<td>29</td>
<td>(\sigma/\sigma_0 = 0.0121 \times N_f^{-0.023})</td>
</tr>
<tr>
<td>Dendorfer et al., 2008</td>
<td>Distal femur – bovine bone</td>
<td>30</td>
<td>(\sigma/\sigma_0 = 0.0133 \times N_f^{-0.094})</td>
</tr>
<tr>
<td></td>
<td>Vertebral – human bone</td>
<td>14</td>
<td>(\sigma/\sigma_0 = 0.0098 \times N_f^{-0.109})</td>
</tr>
<tr>
<td></td>
<td>Distal femur – human bone</td>
<td>14</td>
<td>(\sigma/\sigma_0 = 0.0103 \times N_f^{-0.108})</td>
</tr>
<tr>
<td>Present study</td>
<td>Distal femur – bovine bone</td>
<td>24</td>
<td>(\sigma/\sigma_0 = 0.0197 \times N_f^{-0.119})</td>
</tr>
</tbody>
</table>
Some other struts experienced a rather severe circumstance in which the strut failed in a manner similar to the bending effect with striation of fibrils as shown in Fig. 10(c). Buckling in individual struts can also be observed (Fig. 10(d)) with fracture bands starting near the curve of the individual strut and radiating along its body to a weak region where the microcracks accumulated and crumpled. These three distinct regions are due to multi-cycle loading-unloading which can be visible clearly at superior deformation where the torque load is directed and pointed out.

4. Discussion

The overall goal of the present investigation is to understand the effect of torsional loading and its contributions to the trabecular compression fatigue properties. Mechanical loading has been shown to be both beneficial and detrimental to the function of trabecular bone, depending on the nature and frequency of the loading applied. It has been shown that off-axis load shortens the lifespan of bone dramatically, with certain conditions particularly resulting in dire consequences.

Fig. 9 – (a) Surface fractures on a sample under pure compression at 50C after about 1 million cycles, and (b) microcracks on the trabecular strut of the same sample at higher magnification.

Fig. 10 – SEM micrograph of a sample under compression–torsion at 50C25T, with observations on (a) surface fracture, (b) microcracks on struts, (c) fracture on individual strut and (d) buckling effect. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).
(Dendorfer et al., 2008). Despite the known contributions of torsional loading in fatigue fractures of bone, related studies appear to be lacking (Bergmann et al., 2001; Homminga et al., 2002). In the present study, we investigated the effects of torsional loading in combination with compression fatigue which should provide us with the predictive outcomes of fatigue life. It is apparent that torsional forces are the major factor that needs to be considered since these can lead to fatigue fractures. The data presented in this study is corroborated with other studies (Chang et al., 1999; Kopperdahl and Keaveny, 1998; Morgan et al., 2001).

In monotonic tests, torsional yield stress of bovine trabecular was found lower than the compressive yield stress (Table 2). This suggests that lower stress is needed for the onset of microcrack in the sample under torsional loading than the stress needed in compression. Further, Wang et al. (2005) and Wang and Niebur (2006) have evaluated the effect of change in load onto damage behaviour of the trabecula. It was observed that even at low stress level torsional load may induce the progression of microcracking and accelerate failure. While it is difficult to quantify the exact crack propagation and damage mechanism involved, the decrease in fatigue life of the samples under combined fatigue (Fig. 6) as well as the presented fractured surfaces as in Figs. 9 and 10 agree to this notion. In combined fatigue, two parameters are involved – the loading and the orientation – and both are linked and inseparable. Under pure compression fatigue, fatigue life of the sample was recorded the longest, and the fatigue life shortened as the torsional load was superpositioned and increased. Here, the failure of the sample is said to be ‘torsional dominant’. Furthermore, the maximum strain recorded for samples at different compressive level with similar torsional level demonstrated no significant difference. On the other hand, a high level of torsion increased in compressive levels did not affect both the maximum strain and cycle to failure. Thus, compressive loading is seen as a minor contributing component involved in samples destruction.

Fatigue behaviour of the trabecula could be predicted accurately under multiaxial loading regime as the trabecular fractures in vivo are associated with shear stresses (Greenspan et al., 1998; Pinilla et al., 1996). Thus, results from this study are complementary to previous reported data in the literature. The relationship of normalised stress exponential power with the number of cycles recorded in the present work is in agreement with previously published studies (Table 4). An endurance limit was at approximately 0.0045 of normalised stress level in the case of absence fatigue failure. This value corresponds to the reported fatigue strength at about 0.005 of normalised stress (Moore et al., 2004) and the reported damage threshold of 4000 μs (Moore and Gibson, 2003) for bovine tibial bone.

Variance in the interpolated value at cycle 10^5 as shown in Fig. 7 is associated with the difference in applied loading rate, loading mode and determination of initial stiffness. Thus, difference in fracture mechanism of the trabecula under monotonic is expected in comparison to that under fatigue. In general, the fatigue behaviour of bovine trabecular bone begins when plastic deformation is initiated, followed by strain accumulation and modulus reduction. Subsequently, the strain rate is increased rapidly and more energy is dissipated before the sample finally fails. This behaviour is also typically observed in metallic foams (Gibson, 2000). Understanding the evolution of modulus and maximum strain may provide better insight into the underlying fatigue mechanism of the trabecular failure. Maximum strain is linearly related to the normalised modulus and residual strain (Moore and Gibson, 2003). It is worthy of note that the maximum strain accumulation observed in samples under combined fatigue is higher than that observed in compression alone. However, Lindahl (1976) and Mosekilde and Danielsen, (1987) suggested a 6% surplus of failure strain to be deemed unreliable due to the inordinate experimental error. Thus, a trip limit was set at 6% strain on all fatigue tests in order to prevent crushing in samples. The strain at failure is not loading dependent due to variations in the structural properties of samples, i.e. initial modulus and density. Damage accumulation in samples varies by the modulus, with no regard to the loading applied. Loading modes can also account for the reduction in modulus and stiffness of materials (Wang et al., 2005; Wolfram et al., 2011). This is due to the distribution of stress at tissue levels. Rapid accumulation of damage is the highest in torsion as high tensile stresses induce damage on the trabeculae by bending. Thus, stiffness reduction is mostly observed in samples with superpositioned torsional load (Fig. 8), with heterogeneous stress distribution causing bending on the trabeculae (Fig. 10).

The propagation of microcrack is a common mode of damage which occurs in trabecular bone (Wang and Niebur, 2006). Upon falling or other severe injuries, the observed microcracks and failure in trabeculae as depicted by Fig. 10 may be found to exist in vivo. The micrograph of the fractured sample revealed a few more information on the loading modes and how it affects failure. Microcracking is known to have detrimental effects on the mechanical properties of the trabecula (Burr et al., 1998; Moore and Gibson, 2002; Wang et al., 2005; Wang and Niebur, 2006). A few other studies on bone across species have demonstrated the effect of trabecular microcracking on the fracture of the whole bone (Hoshaw et al., 1997; Kummari et al., 2009; Wu et al., 2013). The failed sample under pure compression fatigue demonstrated unidirectional cracks with fibrillar connections in between them, which suggests the microcrack propagation is initiated by the presence of mineral crystals (Fantner et al., 2004). Here, the oblique-like fracture is observed, and it is generally shear and bending induced. Torsional failure starts with spiral fracture line along the sample axis, resulting in sharp points and edges at the fracture surface. This is often associated with movement of the bone while one extremity is stationary. However, in combined fatigue, the trabecular structure is broken as the mineral crystal is damaged and loses its ability to absorb energy. As a consequence, sample under combined fatigue failed in brittle-like manner with crack propagation in multiple directions. As seen in Fig. 10, the microcrack formation can be associated with the reduction of secant modulus seen in Fig. 8. The development of this microcrack is due to the structural deformation of the sample by shear and bending. Damage is observed to accumulate at the weak region of the strut (as shown by the blue band in Fig. 10(d)). Further load cycle is expected to propagate...
the cracking flakes (as shown by the yellow band in Fig. 10(d)) and accumulate in the same manner.

We are not aware of any other study quantifying the effect of torsional loading on the compressive fatigue behaviour of bovine trabecular bone. Most of the work done in the literature involved mechanical analyses of the trabecula under uniaxial loading regime. There are a few strengths of the presented work that are worthy of notice. Precise mechanical testing methods have been implemented in this study to evaluate the mechanical properties of the tested material. Thus, understanding of the torsional loading effects on the mechanical properties of bovine trabecula have been clarified in a broader scope. Furthermore, in vitro testing allows for the evaluation of physiological loading components to be done individually or in combination. In vitro testing also eliminates the effects of bone remodelling, thus enabling loading–fracture relationship to be investigated directly. Here, authors can accurately quantify the torque magnitude required for the lifetime reduction of trabecular bone. Identification of loading components responsible for subsequent damage can also be done. The information provided herein should shed some light on the damage mechanism involved in whole bone injuries as well as in the development of trabecular bone analogous material.

There are a few limitations in the interpretation of results that should be considered. This study has employed trabecular samples from only one anatomic site of bovine femoral bone. As such, the microarchitecture parameters may differ from that of the human bone or other anatomic sites. However, there are a few works done that demonstrate agreement between the architecture as well as mechanical properties of bovine trabecular bone and that of a healthy human (Keaveny et al., 1994; Morgan and Keaveny, 2001). Furthermore, the trabecular bone behaviour under multiaxial fatigue has never been investigated before, thus direct comparison of the results from this study could not be done.

5. Conclusion

The present study demonstrates the torsional loading effects on the compression fatigue properties of bovine trabecular bone. Decrease in the quality of the trabecular properties are observed in terms of stiffness, life and structural integrity. Damage in the trabecular structure depends on the loading modes. However, further evaluation of microdamage may provide better insights into the failure mechanism of the trabecular bone during fatigue. It is hoped that results from this study will improve the understanding of the behaviour of trabecular bone under combined fatigue and help to develop future assessment of trabecular failure.

Acknowledgement

This project was sponsored by the Kementerian Pendidikan Malaysia (KPM) through Grant scheme (R.J130000.7809.4F355). The authors would also like to thank the Research Management Centre, Universiti Teknologi Malaysia, for managing the project. One or more of the authors are supported by the HIR-MOHE research grant initiative.

Appendix A

The torque, \( T \) (Nm) and the rotational displacement, \( \theta \) (rad/m) were obtained during the test (Fig. 11). In general, the curve of the torque is presented as a function of the angular deformation per unit length \( (\theta = \phi/L) \). The gauge length, \( L \) of the sample was measured in exposed length between mounting and end caps. The maximum torque, \( T_{\text{max}} \) and the twisting stiffness, \( k \), is defined as the slope of the initial region of rotational displacement over torsion under pure torsion test. The shear strain, \( \gamma \) (%) is calculated by Eq. (A1), where \( r \) is the radius of sample.

\[
\gamma = \frac{\theta r}{L}
\]  

(A1)

The shear stress was calculated from the apparent-level torque versus angle of twist curve using the following equation (Nadai, 1950) for an isotropic material.

\[
\tau = \frac{1}{2\pi} \left( \frac{dT}{d\theta} + 3T \right)
\]  

(A2)

where \( T \) is the torque, \( r \) is the radius of the specimen and \( \theta \) is the angle of twist per unit length. The term \( \frac{dT}{d\theta} \) was determined by fitting a fifth-degree polynomial to the torque versus angle of twist curve. While this equation has been used in various studies of trabecular bone (Bruyere Garnier et al., 1999; Ford and Keaveny, 1996; Rincón-Kohli and Zysset, 2009), it can be applied directly only to isotropic homogenous continuum materials and works well also for transversely isotropic materials that have an axis of symmetry along the axis of the cylindrical specimen (Ford and Keaveny, 1996). To calculate this derivative, the experimental torque curve is fitted to an exponential expression a(1−bexp(−c\theta)) (Bruyere Garnier et al., 1999). The slope is obtained from the fitted points by a linear regression. The calculated value of \( \frac{dT}{d\theta} \) is used in Eq. (A2) to obtain the shear stress-strain curve. The shear modulus, \( G \) was calculated by the Eq. (A3)

\[
G = \frac{k \times L}{J}
\]  

(A3)

where \( k \) is the torsional stiffness, \( L \) is the gauge length, and \( J \) is the polar moment of inertia for a cylindrical area.

Using Eq. (A2), the maximum shear stress \( \tau_{\text{max}} \) representing shear strength was calculated by substituting the torque value \( T \) with its recorded maximum value \( T_{\text{max}} \) at the peak point of the torque–angular deformation curve where the

![Fig. 11 – Typical torsional deformation curve to obtain torsional modulus and maximum torque.](image)
slope was zero ($dT/d\phi = dT/d\theta = 0$)

$$r_{\text{max}} = \frac{3T_{\text{max}}}{2\pi^3}$$ (A4)

REFERENCES


Pinilla, T., Boardman, K., Bouxein, M., Myers, E., Hayes, W., 1996. Impact direction from a fall influences the failure load of the
proximal femur as much as age-related bone loss. Calcified Tissue Int. 58, 231–235.