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Investigation of the mechanical response and deformation mechanism of cortical bone material under combined compression and bending loads

Xingdong Sun , Ke Xu, Jie Wang, Liangyuan Xu, Liangfei Fang, Rui Jiang and Yingchun Yang

School of Engineering, Anhui Agricultural University, Hefei, People's Republic of China

E-mail: xdsun@ahau.edu.cn

Keywords: mechanical properties, cortical bone, deformation mechanism, combined loads, experiment

Abstract

Objective service load is the load pattern of cortical bone in practical conditions. The objective service load conditions of cortical bone are complicated, usually including two or more load patterns. The mechanical behavior and deformation mechanism of cortical bone material under coupling load pattern and single load pattern are diametrically different. However, nowadays, researches on the mechanical response of cortical bone have been heavily focused on the single load pattern, which couldn't reveal the potential deformation mechanism accurately. For the purpose of obtaining the objective mechanical properties under complicated loading patterns, the mechanical response and deformation mechanism of bone material under compression-bending coupling load were investigated by in-situ test. The research shows that bending strength increased under the compression-bending coupling load than the single bending load. By in-situ observation, the variations of surface strain distribution and cracks directions were the potential reasons for the increase of the bending strength. It was found that the cracks changed from transverse fracture to integrated patterns with transverse fracture and longitudinal fracture. Larger fracture range and tortuous crack propagation increased the fracture energy dissipation, which led to an enlarged bending strength under the compression-bending coupling load. Through theoretical analysis and numerical calculation, the impeded effect to the increasing of bending deflection was dominant before the final fracture with the adding of the compression load. The numerical calculation result was consistent with the result of the experiment. This present work would provide new references to further studies on the mechanical behavior of cortical bone under complicated loading patterns.

1. Introduction

Cortical bone is a biomaterial-that consists of a mineral phase (hydroxyapatite crystals) embedded in an organic matrix (I type collagen fibers). As the main component of human motor system, the cortical bone plays an important role in load transferring, load bearing and organs protection [1]. Cortical bone also has a hierarchical structure, which was crucial to its mechanical properties [2–6]. Under most objective conditions, the loading patterns of cortical bone are complicated, and the deformation mechanisms of cortical bone material under varied load patterns were different [7, 8]. As mentioned above, it was important to investigate the mechanical response and potential deformation mechanism under different loading patterns. For the purpose of understanding the mechanical property of cortical bone under near service loading condition, experiments, numerical simulation and theoretical analysis were carried out. The typical fracture behaviors such as the elastic and plastic mechanical behavior, multi-scale crack and energy dissipation have been extensively studied [9–12].

Cortical bone materials have specific physiological orientation, so that compression is the typical loading pattern. Through uniaxial tension and continuous compression loading, Nyman found that energy dissipation under compression load was achieved by permanent deformation and viscoelastic strain, and it was achieved by

surface energy release under tension loading [13]. It was also found that the complex hierarchical structure was vital to mechanical properties, because the complex internal loads were created based on its hierarchical structure. The direction and magnitude of the deformations are dependent upon the direction and magnitude of the imposed loads [7]. The effect of microstructure characteristics on elastic modulus was investigated using dynamic tension, and the significant correlation was shown between effective elastic modulus and porosity [14]. Therefore, it could be seen that the effects of microstructure characteristics to different directions were different under the same loading pattern. Arjunan have carried out numerous innovative research about the bone implant, such as developing the porous (68%–90%) Ag bone scaffolds with antibacterial properties by the selective laser melting with excellent mechanical behavior for the first time [15]. Considering the reduction of stress shielding, Arjunan also developed a Ti64 sheathed cellular anatomical structure as the tibia implant by Direct Metal Laser Sintering, which could excellently mimic the objective mechanical properties [16]. Meyer observed that poor understanding on the objective mechanical responses under complicated loading modes could result in a low success rate of the bone grafting and an inaccurate design of bone substitute materials [17].

As shown above, although the previous studies provided some useful guidance for understanding the potential deformation mechanism of cortical bone materials, yet they were mostly concentrated on the single loading mode, such as single tension load, single compression load, and single bending load etc Besides, most studies focused on the macro scale. However, the deformation mechanism of bone material in multi-scale in real-time couldn't be obtained by out-situ research. The objective loading mode in practice was complicated, generally included two or more loads, such as compression-bending coupling loads and etc Some published literature has shown that the loading direction of the external load could affect the mechanical reaction of the bone materials and their behaviors in the fracture process [18–20]. Mechanical response of cortical bone under single loading mode couldn't reveal the objective mechanics law [21]. Arjunan considered that the functional classification of biomaterials should undergo the rigorous experimental evaluation, such as safety, mechanical performance and application [22]. Lee found under the physiologic loads, with the age-related bone loss, the trochanter of the femur increased the risk of fracture in a fall [23]. Jakob also observed that researches on mechanical properties of biological hard tissue materials should be extended to micro-scale level and complex loading patterns [24]. Therefore, the purpose of this research is to investigate the mechanical properties and the underlying potential deformation mechanism of cortical bone material under compression-bending coupling load by in-situ testing. This research would provide some new perspectives to the research on mechanical property and failure mechanism of cortical bone under complex loading patterns.

2. Materials and methods

2.1. Sample harvesting

Cortical bone specimens were obtained from the shaft of the femur of the pig being 18 ± 2 months, collected from the slaughterhouse of Charoen Pokphand Group in Changchun, China. All the soft tissues were removed gently with the scraper. Because the longitudinal direction is the loading direction of compression load, as shown in figure 1, a rectangle block was obtained from the shaft of the femur, and the specimens were cut along longitudinal direction with a band saw. In the cutting process, phosphate buffered saline (PBS) was irrigated on the incision [25]. The rectangle block specimens were polished by using emery papers (P1000, P2000, P3000) step by step as coarse grinding. Then the precision grinding was performed with diamond polishing solution. In the polishing process, the surface of the cortical bone specimens was observed by the optical microscope (OLYMPUS BX53M). The specimen with big defects or many defects would be removed. When the surface quality met the requirements, the polishing was finished. At last, the specimens were cleaned by the ultrasonic cleaner (KQ-218, SHUMEI, CHINA) for one hour. Until be used, the specimens were preserved by immersed in PBS at -20°C . To investigate the evolution mechanism of surface strain, the speckle with obvious contrast was sprayed on the specimens. The final specimens with speckle were shown in figure 1. The final dimension was shown in table 1.

2.2. Methods and instrument

In this research, for the purpose of obtaining the mechanical evolution and deformation mechanism of cortical bone material under the single compression load and compression-bending coupling load, the contrast experiments were designed. To obtain the surface morphology continuously, in-situ monitoring technology was used. The experimental group was compression-bending coupling load group, and the control group was the bending load group. For obtaining the deformation clearly with quasi-static way, the compression loading rate was 0.01 mm s^{-1} , and bending loading rate was 0.01 mm s^{-1} . Considering the compression load was a common load, it was expressed by the form of preloading. Therefore, the compression-bending coupling load was achieved by applying pre-compression load and applying the bending load sequentially. The compression load

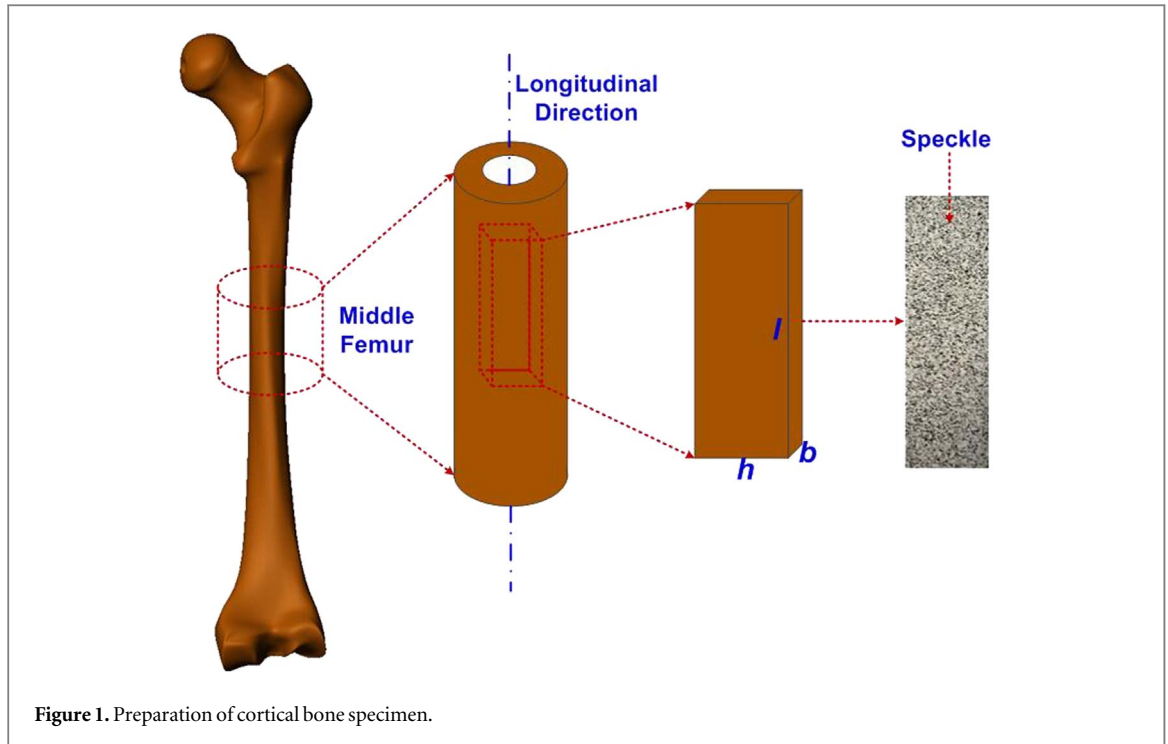


Figure 1. Preparation of cortical bone specimen.

Table 1. The dimensions of the cortical bone specimen (mean $\pm$ standard deviation).

Material	Length(h) (mm)	Width(b) (mm)	Height(l)(mm)
Cortical bone	10.00 ± 0.15	3.00 ± 0.10	50.00 ± 0.30

was kept constant during the loading of bending load. To thoroughly investigate the effect of different pre-compression loads to bending mechanical properties, and to clarify the mechanism differences between combined loading pattern and single bending load pattern, the compression load was set as 500N, 1000N, 1500N, respectively. The experiment method was designed based on the industry standard of mechanical testing of solid materials (JB/T 13221–2017: In situ mechanical property testing system for solid materials with electric thermal magnetic coupling physical field).

As shown in figure 2, a multiple loads material mechanical in-situ testing system was developed, including loading module, signal collecting module, and in-situ observation module. The loading module could provide single compression load, bending load, and compression-bending coupling load. The signal collecting module included load sensor and displacement sensor, and could obtain the load and displacement in the experimental process. The specifications of the signal collecting module are shown in table 2.

In-situ observation module included a digital speckle strain measurement analysis unit, with the function of monitoring and gathering the surface strain of bone specimen. In this experiment, the digital speckle strain measurement analysis unit is ARAMIS system, which is provided by DOM 3D Ltd. The camera resolution is 4096×3000 , and the data registration frequency is 25Hz. The basic principle is as follows. The speckle on the surface of the specimen moves with the deformation of the specimen. By analyzing the speckle patterns before and after the deformation, the relative displacement and deformation of the speckle along the U and V directions (i.e. transverse and longitudinal) are obtained [26]. The calculation method is as follows. Let (x, y) be the point before deformation and (x^*, y^*) be the response point after deformation, and the relationship between them is as follows.

$$\begin{cases} x^* = x + u + \Delta x \frac{\partial u}{\partial x} + \Delta y \frac{\partial u}{\partial y} \\ y^* = y + v + \Delta x \frac{\partial v}{\partial x} + \Delta y \frac{\partial v}{\partial y} \end{cases} \quad (1)$$

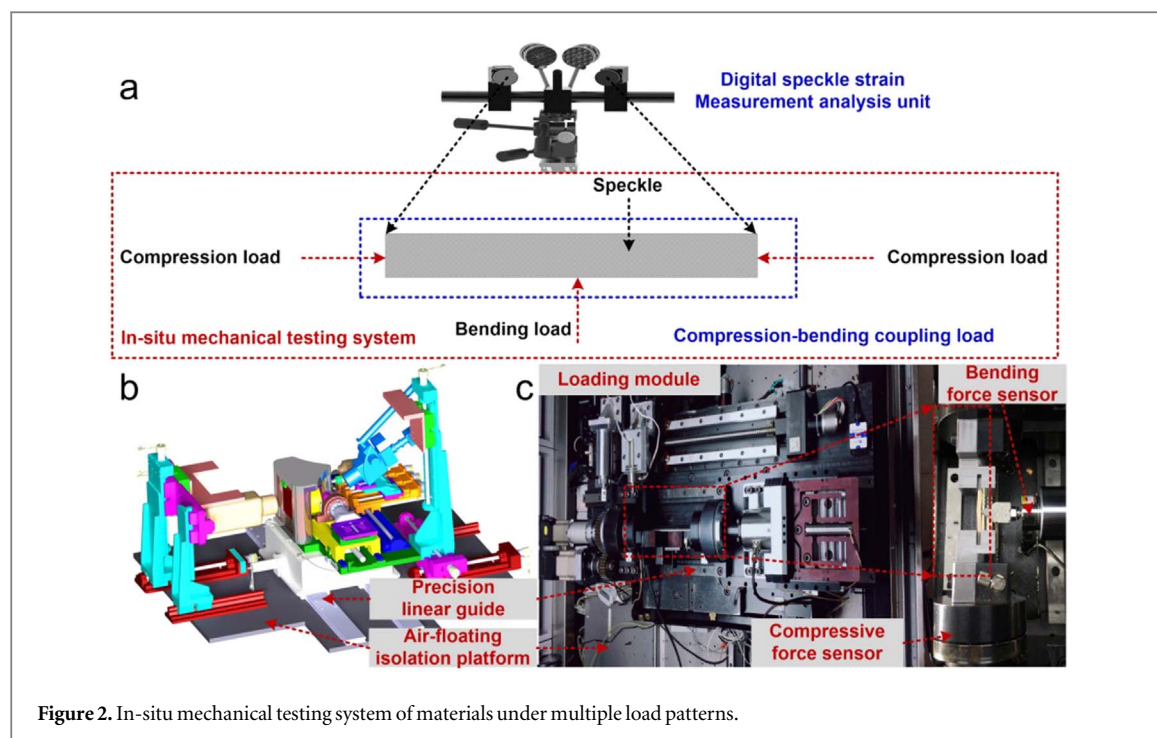


Figure 2. In-situ mechanical testing system of materials under multiple load patterns.

Table 2. The performance of the signal collecting module of the instrument.

Project	Compression	Bending
Load Sensor	Naichuang FM70	CHCCONTACT CF40603
Performance parameter (Maximum value, Resolution)	10 KN, 0.001N	1KN, 0.001N
Loading speed	$1 \mu\text{m s}^{-3} \text{mm s}^{-1}$	$1 \mu\text{m s}^{-3} \text{mm s}^{-1}$
Linear strain	Renishaw RELA/RESA	
Collection data	Compression load	Bending load

To reduce the impact of external vibration, the experiments were carried out on the air-floating isolation platform. Before the experiments, the in-situ testing system was appropriately calibrated.

In this research, to investigate the effect of compression load on the bending behavior, the bending strength and fracture energy of the specimens were chosen as the evaluation indicators. Through contrastive analysis of bending strength under coupled compression-bending load and single bending load, the effect was investigated. The Bending strength is the maximum value of bending stress, and the fracture energy is the consumed energy of the cortical bone specimen until be broken. The bending stress and fracture energy were calculated according to equations (2)–(5) [27].

$$\sigma_b = \frac{M}{W} \quad (2)$$

$$M = \frac{FL}{4} \quad (3)$$

$$W = \frac{bh^2}{6} \quad (4)$$

$$J = \int_0^{t_m} \sigma_b dt \quad (5)$$

In the equations, σ_b is the bending stress, M is the bending moment, W is the interface bending coefficient, F is the bending load, L is the length of the specimen between two supporting point, b is the width of rectangular section, h is the height of rectangular section, J is the fracture energy, t_m is the fracture deflection, and t is the bending deflection.

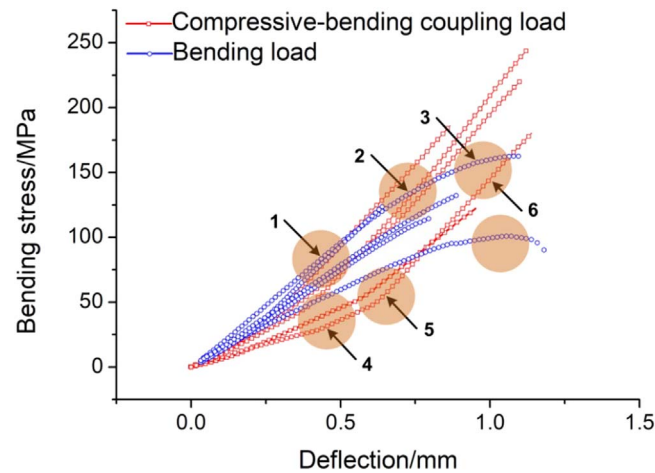


Figure 3. Bending stress-deflection curves under bending load and compression-bending coupling load (Graphs are represented for a single group specimens).

Errors are calculated according to equation (6).

$$s = \sqrt{\left[\sum_{j=1}^n (X_j - \bar{X})^2 \right] / (n - 1)} \quad (6)$$

In this equation, s is standard deviation, X_j is the mechanical property data of each specimen under the same pre-compression load, and $\bar{X}$ is the average value of mechanical property under the same pre-compression load. In this paper, n is 9, and X can express bending strength, fracture energy and deflection value, respectively. There are 36 specimens in all loading patterns.

3. Experimental results and discussion

3.1. Results

As shown in figure 3, the bending stress-deflection curves of cortical bone under different load patterns have been drawn.

The pre-compression-bending coupling load with 1500N compression load was chosen as the control group. The curves have given the relationship of mechanical property between single bending load and compression-bending coupling load. In terms of the ultimate values, it could be observed that ultimate bending stress under compression-bending coupling load was large than that under single bending load in most cases. This could be confirmed in figure 3, where the highest point of curves with red line was higher than that with blue line. The highest point was the fracture point, and the stress beyond this critical point instantaneously decreased to zero. From the trend of the curves, as the bending deflection increased, the increasing rate of the bending stress under single bending load decreased gradually. However, the increasing rate of the bending stress under compression-bending coupling load increased gradually. As shown in figure 3, the black arrows and the brown shadow circles represented the local shape of the curves. It is worth noting that there was some plastic deformation in later stage for the single bending loading pattern. For compression-bending coupling load, there was no obvious plastic deformation until fracture. It could be deduced that the mechanical response was affected by different load patterns. In more details, the compression load along longitudinal direction changed the evolution law of bending stress.

Cortical bone is an anisotropic material, with multi-scale mineralized collagen fibers and different physiological orientations. The specific mechanical response and energy dissipation mechanism under single bending load in different deformation directions are shown in figure 4.

The variations of bending strength and fracture energy with different maximum bending deflection are shown in figure 4(a). It could be observed that the bending strength and fracture energy all increased as the maximum bending deflection decreased. By linear fitting, it was found that the increasing rate of bending strength was less than that of fracture energy. In figure 4(b), a nonlinear relationship is shown between σ_b/J and deflection. The σ_b/J value tends to be smaller, and the decreasing rate decreased gradually. Therefore, it could be speculated that cortical bone materials could absorb more fracture energy, while bending strength could keep increasing steadily under larger bending deflection.

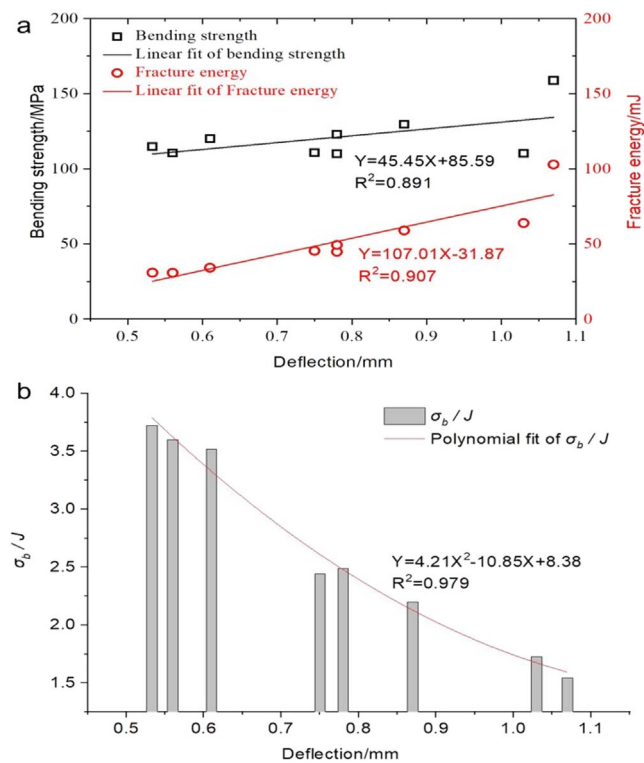


Figure 4. Variations of bending strength and fracture energy with changing of maximum bending deflection (a), σ_b/J -deflection (b).

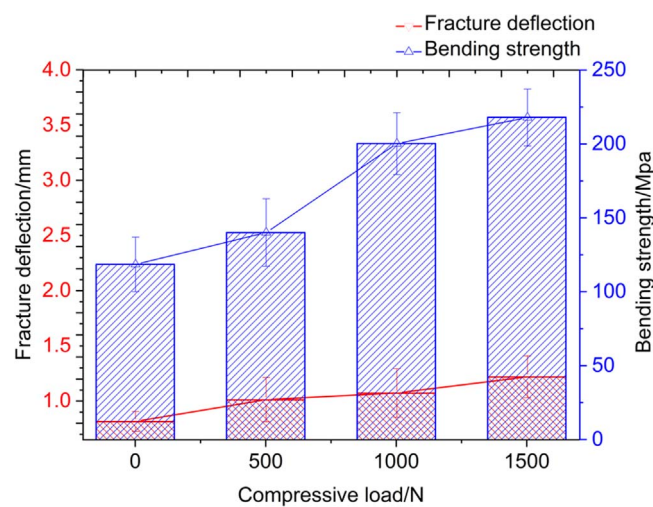
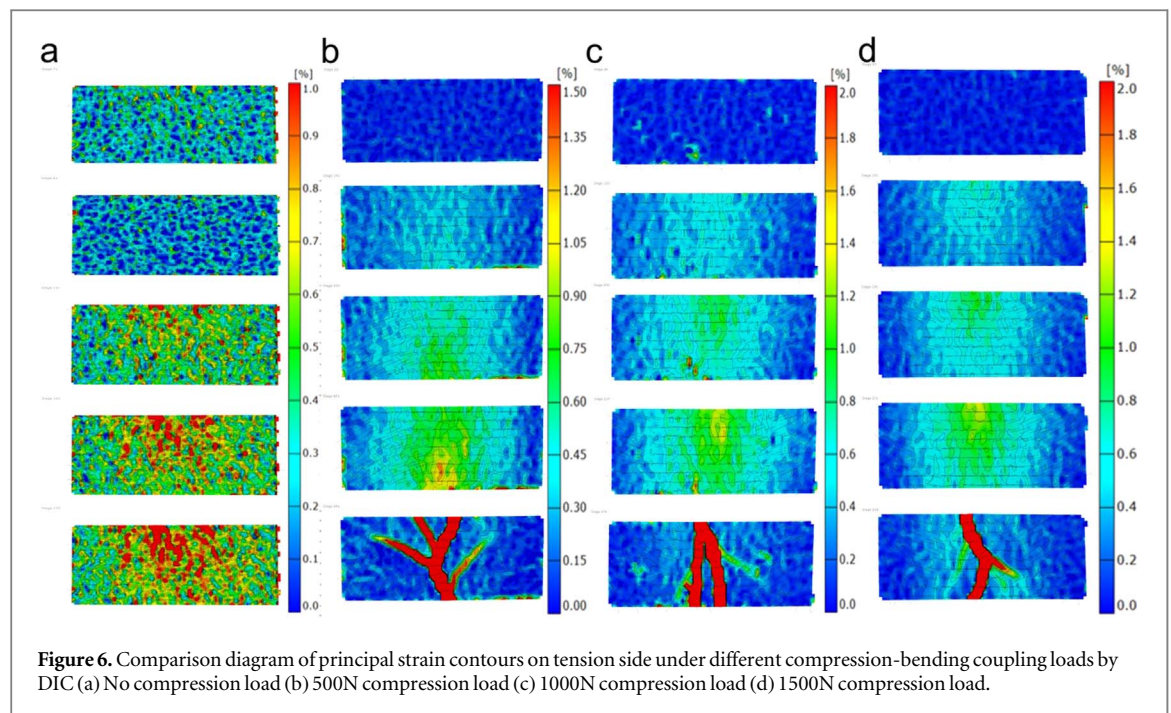


Figure 5. Variations of average fracture deflection and average bending strength under different compression loads.

The variations of average fracture deflection and average bending strength under different compression loads are shown in figure 5. The fracture deflection and bending strength were all the average values of single group with 9 specimens.

The average fracture deflection and average bending strength all increased with the increasing of compression load. When compression load were 500N, 1000N, 1500N, the increasing proportion of fracture deflection were 23.9%, 31.6%, 49.6%, and the increasing proportion of bending strength were 18.2%, 68.9%, 83.9%, respectively. According to the experiments, it could be speculated that compression load could intensify bending strength, and the intensification effect was more obvious with the increasing of the compression load, within a reasonable range.

Fracture in bone material was considered to be especially strain-controlled [28, 29]. To further understand the potential deformation mechanism under different compression loads, the principal strain contours were obtained based on digital image correction method (DIC). The principal strain contours could reveal the



deformation information of the surface area, highly beneficial to intuitively understand the evolution of the mechanical property of bone material. The principal strain contours of the bone specimen's tension side under different load patterns were shown in figure 6. In figure 6, the black line represented direction of principal strain, and the color represented the magnitude of principal strain. Every row of the strain contour represented the different stage from beginning to fracture. In figure 6, the first column is the principal strain contour of cortical bone specimen under single bending load, and the others are the principal strain contours under the combined loads with different compression load.

As shown in eps1 of figure 6(a), under bending load, some sporadic strain concentration micro-zones were generated, with small sizes. In eps2 of figure 6(a), with the increasing of bending stress, the strain increased and the size of micro-zone increased. In eps3 and eps4 of figure 6(a), with further increasing of bending stress, the strain concentration micro-zones started to merge. In eps5 of figure 6(a), the strain concentration bands throughout the short axis of the specimen were created, so that the bone specimen fractured. As shown in eps2 and eps3 of figures 6(b)–(d), in the middle area, there was a strain concentration with obvious gradient. Some branches were derived from the strain concentration band. Finally, as shown in eps5 of figures 6(b)–(d), once any branch extended through the short axis, the bone specimen was broken. Based on the discussions above, it could be observed that the nucleation, growing, merging and developing of the principal strain concentration area varied from free distribution to gradient distribution under the compression load. As was well known, the anisotropy property plays a vital role in mechanical response of bone material. Therefore, from the evolution of the principal strain, it could be speculated that the original anisotropy property was broken and the new anisotropy property was established by the compression load. The branches of the principal strain contours were more regular than the previous case, so that more energy was needed to weaken the original anisotropy.

For the further investigation, the multi-scale fracture morphology of the cortical bone specimens were obtained and shown in figure 7 (macro-scale) and figure 8 (micro-scale).

In figure 7(a), as red dashed line indicated, the fracture edge presented almost straight line. In figures 7(b)–(d), the fracture edge presented different circuitous trend. Compared to straight fracture, the circuitous fracture needs more external energy. At the same time, as the compression load increased, the longitudinal component of the fracture was increasingly obvious. Even when the compression load was 1500N, the stacking fault feature along longitudinal axis was shown. As mentioned above, under the compression load, the fracture edge tended to circuitous, and the fracture was more difficult to achieve. This was consistent with figure 4(a). Therefore, the bending strength increased with the adding of compression load.

The fracture in microscale was important to reveal the potential deformation mechanism of the cortical bone materials. As shown in figure 8, the micro-scale fracture morphology of cortical bone under different load patterns was obtained based on field emission scanning electron microscope (FE-SEM). To further understand the effect of each single load of the coupling loads, micro-scale fracture morphology under single compression load (figure 8(a)) and single bending load (figure 8(b)) was all analyzed. In figure 8(a), the fracture surface was smooth in total, with no debris in any areas. As is well known, the Haversian system was distributed along

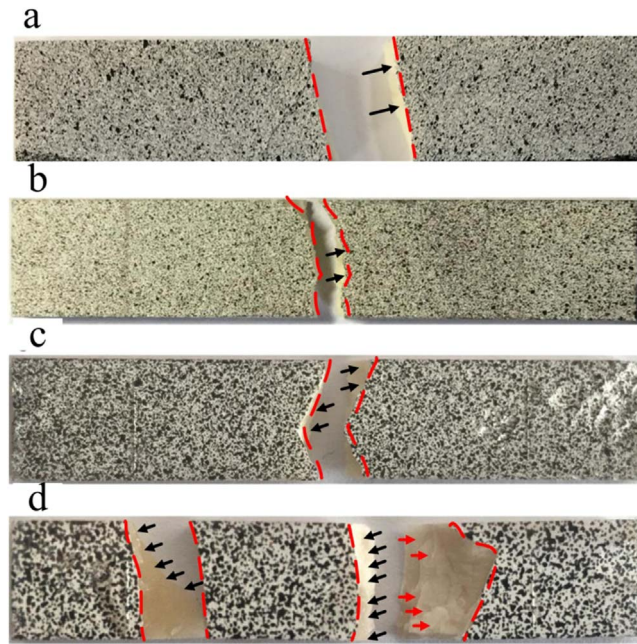


Figure 7. Comparison diagram of macroscopic fracture appearance of the cortical bone specimens (a) No compression load (b) 500N compression load (c) 1000N compression load (d) 1500N compression load.

longitudinal direction. Compared to the fracture feature, it could be observed the fracture direction was the same as the Haversian system orientation under the compression load.

In figure 8(b), the transverse fracture feature was shown, with the rough appearance under bending load. Because the Haversian system was a layered structure, with several layers bone lamellar, and the Haversian system was along the longitudinal direction, the transverse fracture was rougher than the longitudinal fracture. In figures 8(c)–(e), there was grooved zone in the middle area of fracture under compression-bending coupling load. The grooved zones were all along longitudinal direction. It could be observed that the fracture under compression-bending coupling load was the orthogonal pattern, including transverse (perpendicular to Haversian system) fracture and longitudinal (parallel to Haversian system) fracture. Because the fracture area under the compression-bending coupling load was larger than that under the single bending load, the energy consumption was also larger. Finally, the bending strength increased, but the fracture degree was also larger.

3.2. Discussion

As mentioned above, it was observed that the mechanical response and multi-scale deformation mechanism of bone material was influenced under compression load, by variations of the surface strain distribution, crack and fracture pattern. Under the compression load, the circuitous fracture edge and orthogonal fracture pattern was presented, with the larger fracture energy consumption. Finally, the bending strength increased. To further investigate the potential mechanism theoretically, the effects of each load to bending properties (bending strength and bending deflection) were analyzed. As shown in figure 9, the mechanical model of the compression-bending coupling load is established.

As shown in figure 9, it could be observed the bending moment M_1 , from component force F_1 , could promote the bending deflection. An opposite effect was shown on moment M_2 . Both M_1 and M_2 all acted on point C. Based on the analysis above, the coordination effects of promotion and inhibition were the potential reasons for the variations of bending effects. However, the integration effects kept unknown [30, 31].

The hypothesis was proposed that the deformation of the compression side AB and the tension side O_1O_2 was all continuous and homogeneous. In other words, surfaces AB and O_1O_2 were assumed to be cambered surfaces. Therefore, the bending moment M_1 and M_2 could be calculated according to equation (7), (8).

$$M_1 = \frac{2F_C L^2 k (L^2 - 4k^2) - 16F_C L^2 b k^2}{(L^2 + 4k^2)^2} \quad (7)$$

$$M_2 = \frac{F_C (L^2 - 4k^2) (2L^2 k + bL^2 - 4bk^2)}{(L^2 + 4k^2)^2} \quad (8)$$

In the equations, F_C is the compression load, L is the length of specimen, α is the inclination angle of terminal, b is the thickness of specimen, k is the bending deflection.

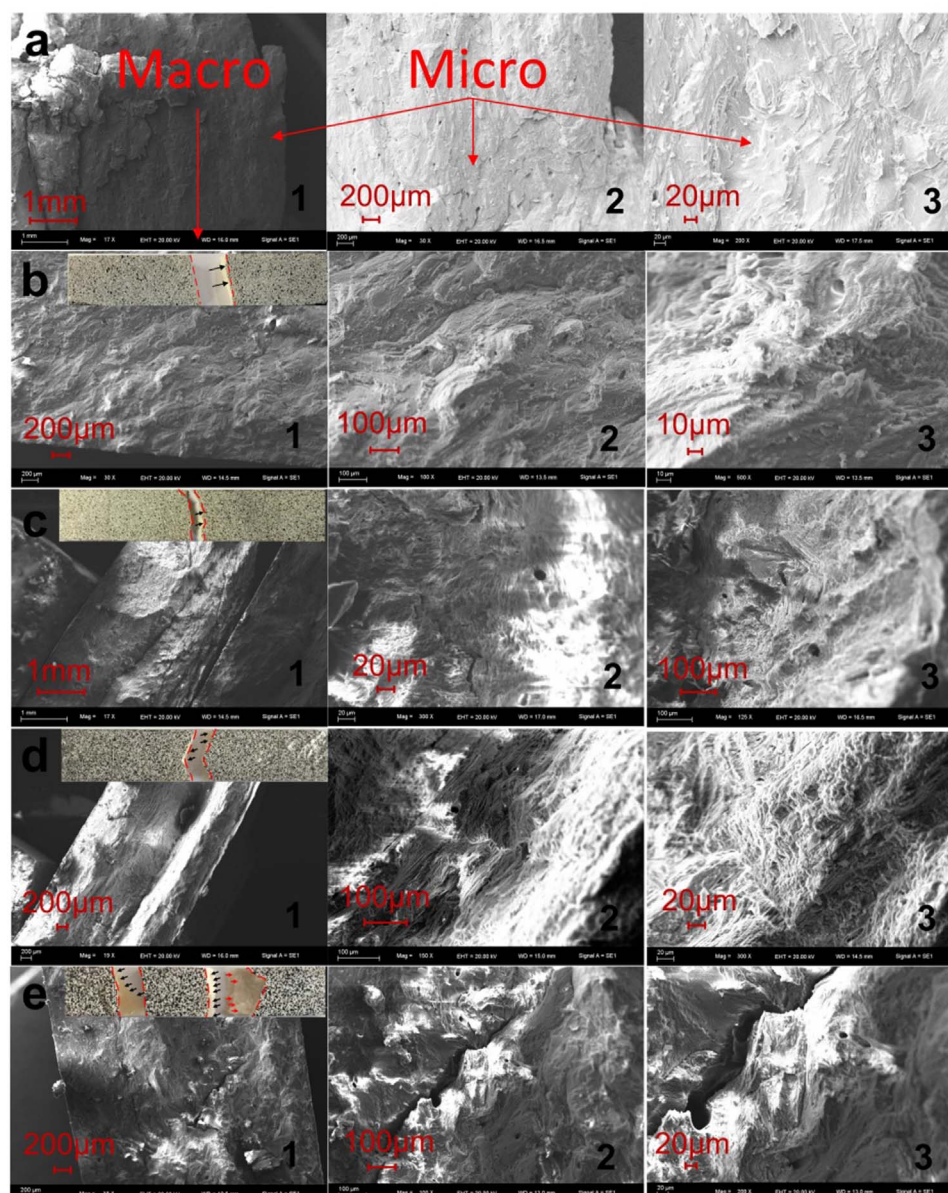


Figure 8. Comparison diagram of microcosmic fracture appearance of specimens (a) Compression load (b) Bending load (c) 500N compression load (d) 1000N compression load (e) 1500N compression load.

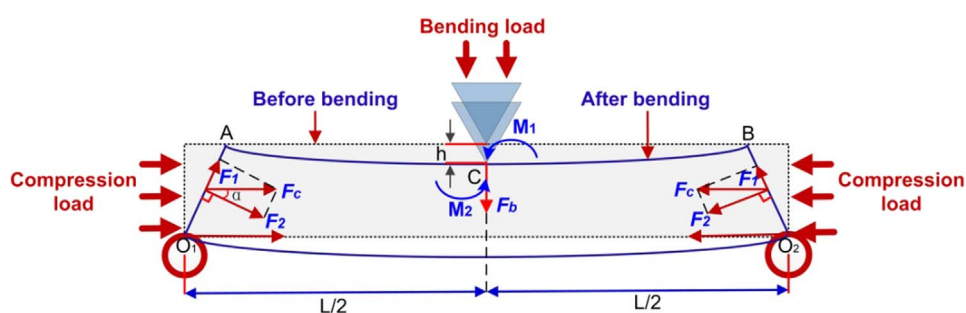


Figure 9. Mechanical model of cortical bone under compression-bending coupling load.

The numerical calculation was carried out, and the bending moment-deflection ($M-k$) curves are shown in figure 10.

It could be observed from figure 10 that the bending moment M_1 increased at first and decreased then, with the increasing of the bending deflection. Bending moment M_2 always decreased with the increasing of bending

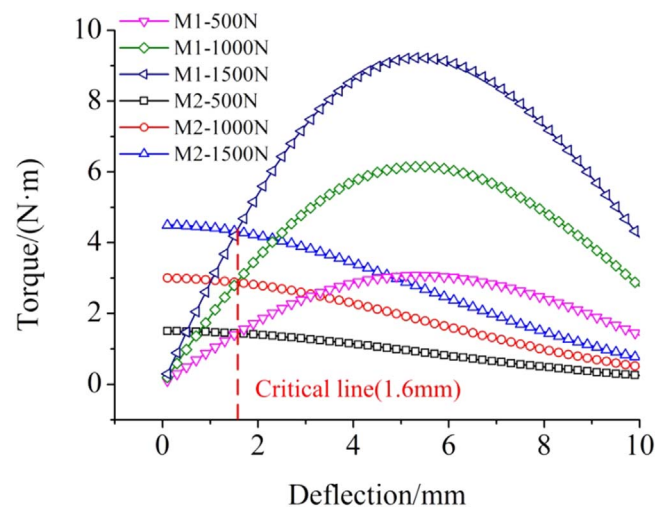


Figure 10. Bending moment-deflection ($M-k$) curves.

deflection. By contrast, the initial value of M_2 was bigger than that of M_1 . When bending deflection was over the threshold value, M_1 was bigger than M_2 . As mentioned in the mechanical analysis, the effect of bending moment M_1 and M_2 to bending deflection was promotion and inhibition, respectively. Therefore, it could be seen that the increasing of bending deflection was impeded at first and promoted subsequently. Based on the analysis above, as shown in figure 3, the fracture deflections of all the specimens were less than the critical value 1.6 mm in figure 10. Therefore, the bending moment M_2 was always bigger than the bending moment M_1 until the final fracture. The inhibition effect was exhibited throughout the whole process of fracture, so that it was protected from invasion of external load. In order to overcome this inhibition effect, a larger bending load was required. Therefore, the bending strength increased under the compression load, as shown in figure 5. The obtained results in figure 10 were consistent with descriptions of the bending stress-deflection curves in figure 3 and variations of average bending strength in figure 5.

Cortical bone fracture was the result of strain redistribution, and the bending strength of the bone material was considered to mainly depend on its compression strength [20]. In this research, the compression extents of both tension and compression surfaces under the compression load were all increased, and the strain redistribution of the tension surface implied that the neutral axis of the bone specimen moved towards the compression surface. In the inelastic regime, the volume of cortical bone under compression load was nearly constant. Ebacher also considered that tension strain rate increased faster than its compression strain rate [20]. Therefore, the strain redistribution of the tension surface might be much heavier. It was well known the compression strength of bone matrix was bigger than its tension strength, so the micro-crack initiated from the tension surface firstly. Haversian system served as the energy absorber to prevent and delay the crack propagation [32]. Therefore, adding the compression load, the bigger elongation in tension surface was achieved until final fracture, so that it resulted in a greater damage and a larger bending strength. It was consistent with figure 7.

In terms of the mechanical response under the bending load, Currey observed that post-yield deformation was one of the main factors to determine the bending strength [33]. As shown in figure 3, it could be seen that the post-yield deformation under the single bending load was obvious. There was no obvious post-yield deformation under compression-bending coupling load. It could be some correlations with the larger damage range under compression-bending coupling load. As was well known, the inelastic strains occurring in bone were mainly associated with micro-damage [34–36]. The micro-damage morphologies in both compression surface and tension surface were totally different [37], because of the different carrying capacity of the collagen fibers and hydroxyapatite microcrystal under the compression strain and tension strain [38]. In this research, compared to single bending load, the cortical bone specimens under the compression-bending coupling load have been shortened previously, resulting in a reduced size of the collagen fibers and hydroxyapatite microcrystal in the long axis direction. The micro-porosity between the mineralized collagen fibers staggered structure decreased with the matrix shrinkage effect. Therefore, the increase of deformation was growingly difficult. To overcome the bigger binding force in the microstructure of bone material, the bigger deflection and bending strength were needed than that under the single bending load pattern. It was consistent with the descriptions in figure 3.

Trebackz considered that the anisotropy was weakened under the compression load because of the shifting and friction between collagen lamellae and their gradual separation [39]. It was consistent with the descriptions in figure 6. Although the low anisotropy could result in a decreased mechanical property, the restricted connections at both ends hindered the fracture. At the same time, the strain concentration in the middle area before the final fracture was bigger than that under single bending pattern. Therefore, the fracture area and range were enlarged. It was consistent with the variation of anisotropy of the cortical bone in figures 7 and 8. It could be seen that the compression load weakened the anisotropy of cortical bone, but also provided a resistance to the disadvantage from low anisotropy. It was also consistent with the research of Currey, the cortical bone could still carry load even after large deformation under the compression load [40]. In this perspective, the compression load influenced the bending behaviors by changing the anisotropy of the cortical bone material.

In a homogenous material, the straight crack propagation needs less energy dissipation, whereas in a heterogeneous material, the tortuous crack propagation needs more energy [41, 42]. Therefore, as a natural heterogeneity biomaterial [38, 43], the tortuous crack propagation under the compression-bending coupling load was the potential reason for the increased bending strength [20]. Based on the discussions above, the research findings could be mutually confirmed with the previous studies.

4. Conclusions

In this paper, the mechanical response of cortical bone under the compression-bending coupling load was investigated by in-situ experiment. The deformation mechanism was discussed from the strain evolution and multi-scale structure of bone material. Conclusions are drawn as follows:

- (1) Our data indicated that the bending strength increased under compression-bending coupling load than single bending load. Compared to the single bending load, the surface principal strain presented a more significant gradient under the compression-bending load. The anisotropy of the cortical bone specimen decreased under the compression-bending coupling load than single bending load.
- (2) The compression load weakened the anisotropy of the bone material, but also provided a slow-release to the disadvantage from the low anisotropy. The double-edge effect was shown about the compression load, which caused the ultimate bigger bending strength and also the bigger disruption. There was a critical deflection between the protective effect and destructive effect of compression load.
- (3) The objective service mechanical properties of cortical bone should be investigated deeper considering the shape of the whole bone and the muscle wrapped around the bone. For future research, it is necessary to create a complicated bone-muscular system and investigate the mechanical response of the whole system. This would bring more beneficial discoveries. This research would provide the theoretical references for the mechanical reliability evaluation of the bone substitute materials, and also provide the reference evidences for developing of the avoidance strategy on the dangerous stress conditions and for identifying of different damage conditions.

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Data availability statement

The data that support the findings of this study are available upon reasonable request from the authors.

ORCID iDs

Xingdong Sun  <https://orcid.org/0000-0002-1781-3599>

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