Journal of the Mechanical Behavior of Biomedical Materials 77 (2018) 192–198

Contents lists available at ScienceDirect

Journal of the Mechanical Behavior of Biomedical Materials

journal homepage: www.elsevier.com/locate/jmbbm

Experimental study of time response of bending deformation of bone cantilevers in an electric field

Huimin Kang[a], Zhende Hou[a,*], Qing-Hua Qin[b]

[a] Department of Mechanics, School of Mechanical Engineering, Tianjin University, Tianjin 300072, China
[b] Research School of Engineering, Australian National University, Acton, ACT 2601, Australia

ARTICLE INFO

Keywords:
Bone
Electromechanical coupling
Electrostriction bending deflection
Polarization

ABSTRACT

Bone is a complex composite material with hierarchical structures and anisotropic mechanical properties. Bone also processes electromechanical properties, such as piezoelectricity and streaming potentials, which termed as stress generated potentials. Furthermore, the electrostrictive effect and flexoelectric effect can also affect electromechanical properties of the bone. In the present work, time responses of bending deflections of bone cantilever in an external electric field are measured experimentally to investigate bone's electromechanical behavior. It is found that, when subjected to a square waveform electric field, a bone cantilever specimen begins to bend and its deflection increases gradually to a peak value. Then, the deflection begins to decrease gradually during the period of constant voltage. To analyze the reasons of the bending response of bone, additional experiments were performed. Experimental results obtained show the following two features. The first one is that the electric polarization, induced in bone by an electric field, is due to the Maxwell–Wagner polarization mechanism that the polarization rate is relatively slow, which leads to the electric field force acted on a bone specimen increase gradually and then its bending deflections increase gradually. The second one is that the flexoelectric polarization effect that resists the electric force to decrease and then leads to the bending deflection of a bone cantilever decrease gradually. It is concluded that the first aspect refers to the organic collagens decreasing the electric polarization rate of the bone, and the second one to the inorganic component influencing the bone's polarization intensity.

1. Introduction

Bone is a kind of hard tissue that plays a critical role in supporting the whole body and maintaining the life activity of a human being. The main functions of the bone are to bear stresses and to support the weight of human bodies. In turn, external stresses can alter the structure, shape, and density of the bone to adapt to the load, known as bone remodeling, which obeys Wolff's Law. Moreover, bone has unique electromechanical behaviors, such as piezoelectric effect (Fukada and Yasuda, 1957; Qin and Ye, 2004; Qu et al., 2006) and streaming potentials (Anderson and Eriksson, 1968, 1970; Pienkowski and Pollack, 1983), which are termed as stress generated potentials in bone (SGP).

Bone is complex not only in its hierarchical structure with anisotropic mechanical properties but also in its electromechanical properties. Various methods have been developed for comprehensive understanding of the electromechanical properties of bone (Atsushi, 2015; Hastings and Mahmud, 1988; Isaacson and Bloebaum, 2010; Qin et al., 2005; Qin and Ye, 2004; Ren et al., 2015; Rosa et al., 2015). It was found that when a bone is being loaded, its piezo-voltage decay follows a stretched exponential law (Hou et al., 2011). Using a piezoelectric force microscope, Halperin et al. studied the piezoelectric effects in both, wet and dry bone at nano-scale and obtained a piezo-response image with nanometer scale resolution (Halperin et al., 2004). Aschero et al. investigated the converse piezoelectric effect of bone through measuring its bulk change induced by an electric field (Aschero et al., 1996). Wieland et al. used X-ray micro-diffraction to study the inverse piezoelectric effect of bone and measured the shear strain induced in a femur by an electric field (Wieland et al., 2015). Itoh et al. reported that charges induced by an external electric field can affect bone growth as well as osteoblast activity (Itoh et al., 2006), which indicated that electrical signals may play an important role in bone remodeling processes.

Studying the electromechanical properties of bone not only can help us to understand the nature of bone materials, but also have clinical significance. This study indicates that a square waveform electric field can makes a bone cantilever bend and the bending response of the bone with time is associated with the collagens in bone. Additional experiments reveal several aspects of the electromechanical properties of

* Correspondence to: Department of Mechanics, School of Mechanical Engineering, Tianjin University, No. 92 Weijin Road, Nankai District, Tianjin 300072, China.
E-mail address: hou@tju.edu.cn (Z. Hou).

http://dx.doi.org/10.1016/j.jmbbm.2017.09.017
Received 25 May 2017; Received in revised form 27 July 2017; Accepted 11 September 2017
Available online 13 September 2017
1751-6161/

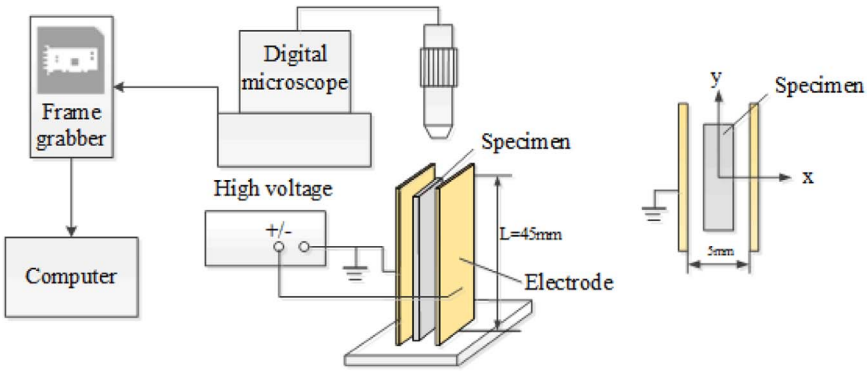

Fig. 1. Measurement system and top view of a specimen and the electrodes.

bone.

2. Materials and methods

2.1. Specimen preparation

Seven cortical bone specimens used in this study were taken from the mid diaphysis of dry degreased bovine tibias (age 2–3 years) and machined into rectangular beams with a length of 60 mm, thickness of 0.8 mm and width ranging from 7 to 8 mm. Each specimen's axes and surfaces were parallel to the axis and side surface of the diaphysis respectively.

2.2. Experimental setup

The measurement system is illustrated in Fig. 1. The specimen was clamped at its bottom end in a cantilever fashion and was placed between a pair of parallel copper plate electrodes 5 mm apart. The span of the cantilever was 45 mm. A high voltage amplifier (Trek 610D-k-CE H.V. Supply Amplifier/controller, Trek Inc.) was used to apply a voltage between the electrodes, producing a uniform electric field in the gap between them. The left electrode was grounded and the right one was connected to the voltage output terminal. The direction of the electric field could be changed by a switch on the instrument panel. A digital microscope (Hirox KH-7700, Hirox Co., Ltd.) was used to take digital images of the upper surface at the free end of the cantilever, which the total time required for capturing, converting and saving one digital image was one second. The size of the view field of the microscope was 110 μm × 82.5 μm corresponding to 1600 pixels × 1200 pixels (resolution: 0.0688 μm/pixel).

In order to investigate the time response of bending deformation of the bone specimens, the measurement system was improved by increasing its image acquisition speed. Taking advantage of the digital microscope having a VGA (Video Graphic Array) output port, a frame grabber (Mafite M1202, Liangrumei Technology Co., Ltd.) was used to convert the video signals, output from the microscope, into digital images synchronously and to save them in the computer. With this grabber the image acquisition speed reached 15 fps, as shown in Fig. 1.

2.3. Experimental procedure

A reference image was taken before an electric field was applied to the specimen. Then, a square waveform voltage with an amplitude of 3000 V was applied between the two electrodes, with the corresponding electric field intensity of 6×10^5 V/m. The waveform of the voltage was recorded by an oscilloscope (LeCroyWavesurfer 3054, Teledyne LeCroy Inc.). The duration of the waveform was about 10 s and the rising and falling edges were shorter than 0.2 ms. Once the voltage was applied, the microscope with the image grabber began to capture the deformed images at an acquisition rate of 15 fps for about 11 s. Total 150–170 deformed images were recorded for each measurement. Then, the details of the bending deflections at the free end of a specimen versus time were obtained by a correlation algorithm (Bing et al., 2009; Sutton et al., 1986) between the reference image and the deformed images.

3. Characteristics of the results

3.1. The bending feature of the specimen

In our previous work (Xu et al., 2015b), we found that a bone cantilever can be bent by the electric attractive force caused by an electric field or voltage. The deflection of the cantilever is proportional to the square of applied voltage, and its bending direction is towards the ungrounded electrode (right electrode) regardless of the sign of the applied voltages. When the right electrode is grounded, the specimen will bend towards the left. The results can be obtained from any of the seven specimens tested, and the significant results of this work are the following.

3.2. Time-response of bending of bone with and without collagens

Fig. 2 shows the measurement results of the specimen 1. The red curve is the deflection and the blue one denotes the waveform of the applied voltage. The positive deflections represent the bending direction towards the ungrounded electrode (right). The red curve shows that once the electric field is applied the bone cantilever begins to bend and its deflection increases monotonously to a peak value within about 1.6 s. After that, the deflection begins to decrease gradually during the constant voltage period, and when the electric field is removed, the deflection drops abruptly to zero.

There are two features in the deflection curve. The first one is that

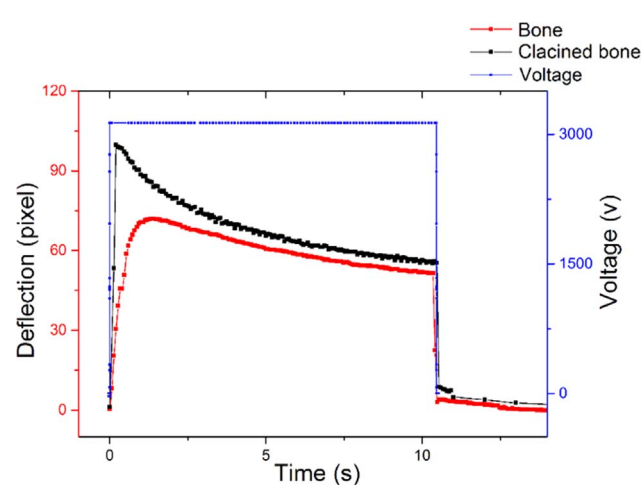

Fig. 2. Deflections versus time of specimen 1 with and without collagens. Red denotes the specimen with collagens, while the black denotes the calcined specimen. Blue denotes the waveform of the applied voltage.

the deflection at the free end of the bone cantilever reaches its peak value in about 1.6 s, which we call delay increase, and the second one is that the deflection decreases gradually during the constant voltage period, seeming like a '*relaxation* '.

Bone is mainly composed of organic collagen and inorganic hydroxyapatite (HA), hence the features of the deflection curve reflect the properties of the collagen and HA. To identify the contribution of the two components to the bending features of the bone, we removed the collagens from the bone specimens, by calcinating them at 500–550 °C for 15 mins in a furnace, which can remove the collagens from a bone specimen completely. The calcined bones become brittle, and their porosity is increased as described in the literatures (Raspanti et al., 1994; Ren et al., 2009; Figueiredo et al., 2010). Literature (Nicholson, 1992) shows that the elastic modulus and modulus of rupture of bone can decrease 80% and 95% respectively at the temperature of 500–600 °C. Five of the seven specimens were calcined successfully, and two were broken into several pieces during calcinating process. Then, bending deflections of the five calcined specimens were measured in the same conditions as described above.

For comparison, the results of the burnt bone (without collagen) and the unburnt bone (with collagen) are plotted in Fig. 2. The red and black curves are the deflection responses of specimen 1 with and without collagens respectively, and the blue curve depicts the waveform of voltage or electric field applied.

The difference between the red and black curves is obvious. Once the electric field is applied, the deflection of bone without collagens increases to its maximum value almost immediately instead of delaying for over one second. However, after the two curves reach their maximums respectively, both decrease gradually in a similar manner, during the constant voltage period.

Seven bone specimens were measured and five of them were calcined and measured successfully. Fig. 3 shows the bending curves of the other four specimens. It is obvious that they have the same bending response as the specimen 1.

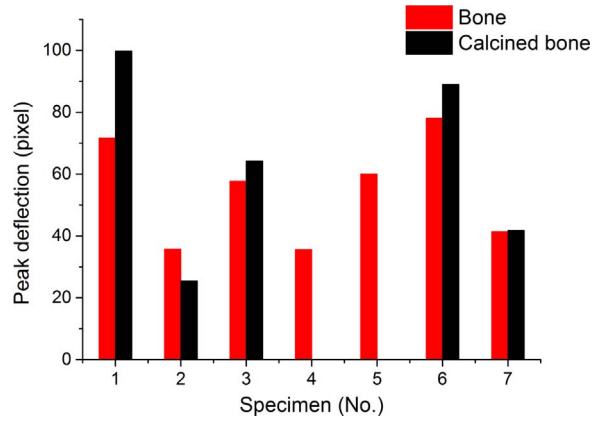

Fig. 4. Maximum deflections of bone and calcined bone.

To identify the time response feature of bending of the specimens, the maximum deflections of the specimens are shown in the column diagram of Fig. 4. The horizontal ordinate denotes specimen numbers. The heights of the red and black columns represent the maximum deflections of the bone specimens with and without collagens respectively. The times corresponding to the peak deflections of bone are listed in Table 1. The peak values of the deflections of the specimens with collagens range from 35.7 to 78.1 pixels (2.12–4.65 μm) and the corresponding times range from 0.73 to 1.67 s. The peak values of the deflections of calcined specimens are from 25.4 to 99.8 pixels (1.51–5.94 μm), and the corresponding times are below 1/15-2/15 s, because the deflection of the calcined specimen increases relatively too fast to be recorded accurately by the image acquisition which is limited in 15 fps.

Because the specimen is in the gap between the two electrodes and does not contact any object, the bending of it is caused by the electric field force which must concern to the electric properties of bone.

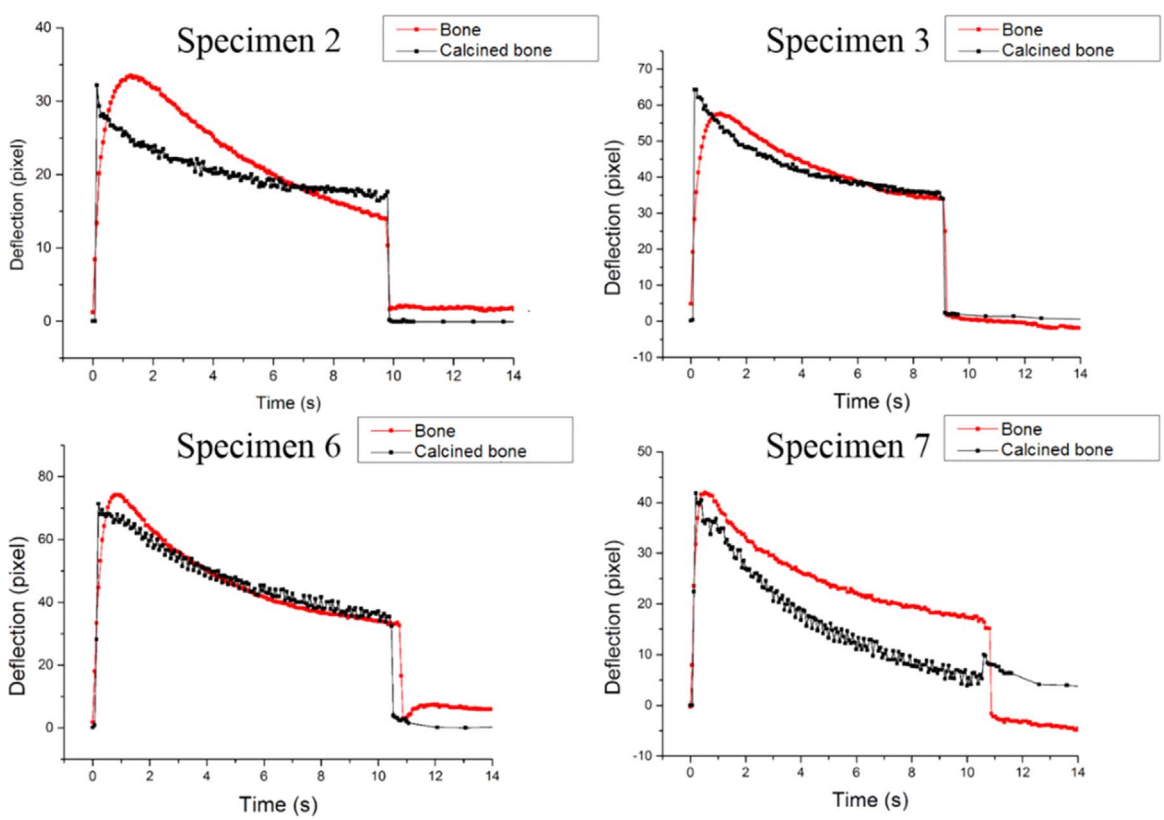

Fig. 3. Deflections versus time of specimen 2,3,6 and 7 with and without collagens.

Table 1
Times corresponding to the deflection peaks.

Specimen Number	1	2	3	4	5	6	7
Time(s)	1.60	1.67	1.13	1.40	1.40	1.20	0.73

However, whether the electric field still affects the mechanical properties of bone is also one of the objectives of following analyses.

4. Discussion

The experimental results demonstrate that the organic component-collagens domain the behavior of delay increase. This is the first result of the study and the consequent issue is to attempt to investigate the electromechanical coupling mechanism, which may include several principles such as piezoelectricity, electrostriction and flexoelectricity or mechanical properties of bone.

4.1. Curve fitting

We fitted the deflection curves using exponential functions and found that each deflection curve of a specimen with collagens can be well fitted by two exponential functions with different time constants. The $x(t)$ below represents the fitting function

$$x(t) = x_0 + Ae^{-t/\tau_1} + Be^{-t/\tau_2} \tag{1}$$

where A, B and $\times_0$ are fitting constants, τ_1 and τ_2 are time constants.

Fig. 5 illustrates the fitting curves for specimen 1, the blue line represents the fitting curve and red scatters are the measured values. All the fitted parameters of the seven specimens are listed in Table 2.

The time responses of the specimens without collagens are also fitted, though they are relatively simple. These curves decrease monotonically and one exponential decay model is employed to fit a curve. Eq. (2) is the fitting function

$$x(t) = x_0 + Ae^{-t/\tau_c} \tag{2}$$

where τ_c is the fitting time constant. Fig. 6 is the fitting curve of specimen 1. Table 3 lists the five fitting time constants of the five specimens without collagens. τ_c is in the same order of magnitude as τ_2 in Eq. (1).

In the fitting function (1), the two exponential terms with different time constants suggest that the form of Eq. (1) is the same as a solution of a motion equation of a vibration system based on a Kelvin viscoelastic model (Flügge, 1975; Nashif et al., 2001), which consists of a spring and a dashpot in parallel. Considering that one end of the model is fixed and a mass m is connected to the other end, and let k represent the spring constant and η as the damping constant. Under the action of

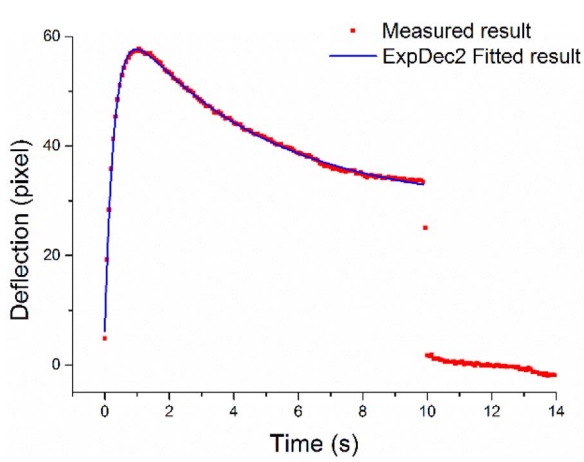

Fig. 5. Fitting curve for specimen 1.

Table 2
Fitting results for specimens 1 to 7 with collagens.

No.	Fitted parameters					Statics	
	x_0	A	τ_1	B	τ_2	Reduced Chi-Sqr	r^2
1	44.931	−83.966	0.382	36.539	6.031	0.40049	0.996
2	5.344	−38.131	0.365	35.375	6.859	0.13523	0.997
3	28.995	−61.33	0.29	38.512	4.334	0.11611	0.998
4	12.615	−45.154	0.619	35.046	5.802	0.11756	0.997
5	30.943	−68.31	0.424	39.998	8.024	1.36026	0.983
6	30.365	−89.142	0.259	59.087	3.608	0.12908	0.999
7	15.351	−49.973	0.156	31.246	3.768	0.57495	0.990

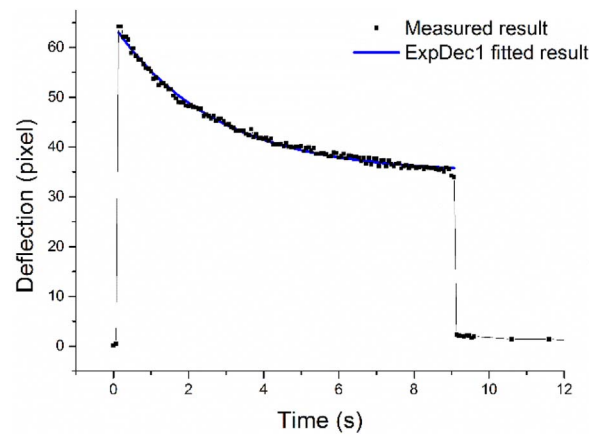

Fig. 6. Fitting curve for calcined specimen 1.

Table 3
Fitting results for the five specimens without organic component.

No.	Fitted parameters			Statics	
	x_0	A	τ_c	Reduced Chi-Sqr	r^2
1	53.779	47.264	3.588	0.642	0.995
2	16.624	12.660	3.314	0.252	0.976
3	34.730	29.779	2.689	0.350	0.994
6	27.668	44.813	5.798	1.626	0.985
7	1.644	40.811	4.436	1.071	0.989

an external force F acting on the mass, the governing equation of the model is

$$x'(t) + \eta x'(t)/m + kx(t)/m = F/m \tag{3}$$

where x is the displacement of the mass from equilibrium position. If the system is over-damped ($\eta^2 - 4km > 0$), and the external force F is a step function, then the solution is

$$x(t) = Ae^{-t/\tau_1} + Be^{-t/\tau_2} + F/k \tag{4}$$

where the time constants τ_1 and τ_2 are the function of the material property parameters η, k, and m. The non-zero constants A and B can be determined by the initial conditions.

If selecting suitable parameters for Eq. (4), it can be equivalent to Eq. (1) in terms of the form and values. However, Eq. (1) does not mean that its bending response just associates with mechanical properties of bone such as elasticity or viscoelasticity. Dry bone is a weak viscoelastic material and does not have over damping properties, which is indisputable. That the deflection dropping to zero abruptly, once the electric field is being removed, displayed in Figs. 2 and 3, demonstrates the strong spring property of bone.

In order to determine whether the mechanical properties of dry bone can be affected by an electric field, an additional experiment was

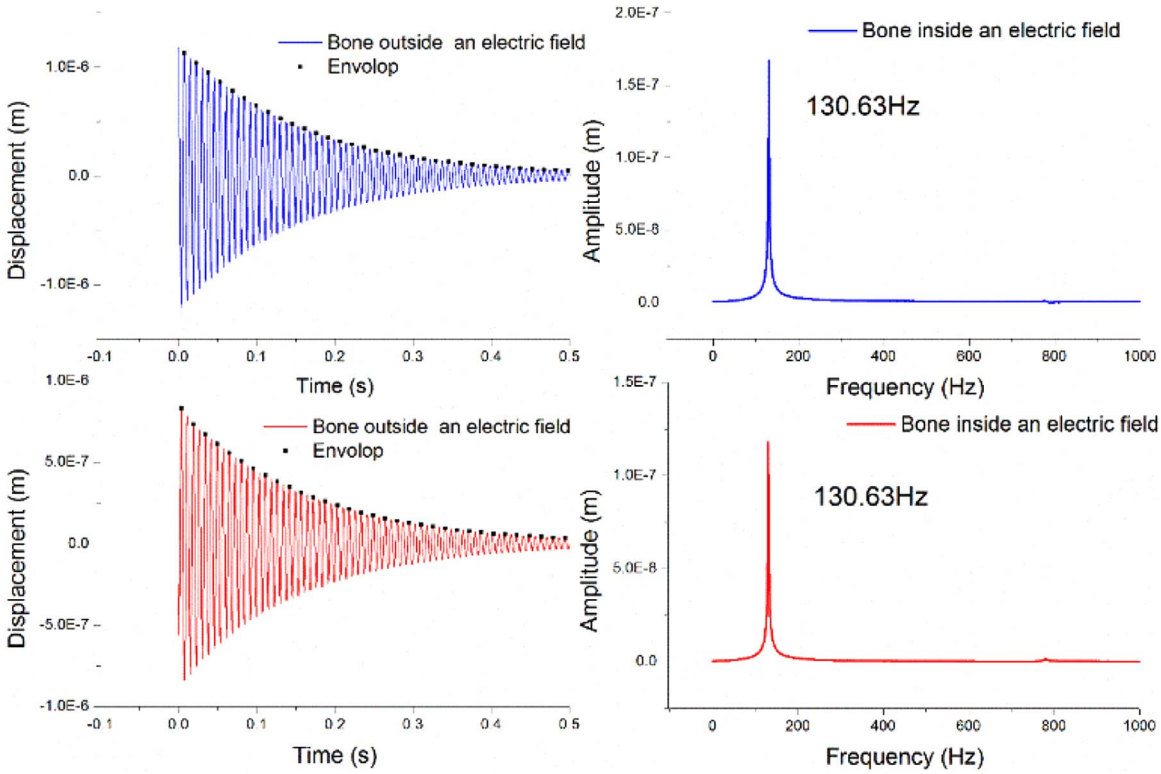

Fig. 7. Displacement vibration and corresponding frequency spectra of bone cantilever inside and outside an electric field.

carried out, for measuring the vibration features of a bone cantilever being inside and outside an electric field respectively. A single-point laser vibrometer (Polytec OFV-5000, Polytec Inc.) was aligned along the *x* direction (Fig. 1), to measure the vibration of the bone cantilever portion over the electrodes. The bone cantilever was excited in free vibration by forcing it a micro bending deflection with a micro-positioning stage.

Fig. 7 shows the displacement vibrations of the cantilever with respect to time and the corresponding frequency spectra. The curves in Fig. 7 show that the cantilever vibrates freely both inside and outside the electric field, which means that dry bone does not have over damping property inside an electric field. The first order natural frequencies of the cantilever in the two states are equal to 130.6 Hz. Generally, the natural frequencies of a cantilever depend on its geometric dimensions, density and elastic modulus. Because the changes of specimen's geometry and density, induced by the electric field, are almost negligible, the equality of the two natural frequencies demonstrates that the elastic modulus of bone in the two states are almost the same.

Therefore, it can be concluded that the features '*delay increase*' and '*relaxation*' reflect the electric properties of bone instead of mechanical properties.

It should be noted that the bending deflection responses of bone, as showed in Fig. 2 and Fig. 3, must include vibration but cannot be measured by the experiment setup shown in Fig. 1. The reasons are that the amplitudes of vibration are too small to be tested by the setup used, and the frame grabber cannot capture high frequency displacement signals.

4.2. Polarization behavior of bone

4.2.1. Maxwell–Wagner polarization in bone

The exponential term with time constant τ_1 (one order of magnitude smaller than τ_2) in Eq. (1) depicts the '*delay increase*', which describes intuitively the main initial increasing portion of the deflection curve. Regarding the physical meaning, it should be associated with electric properties such as dielectric polarization of bone.

The polarization mechanism of bone depends on its structure. Bone is a hierarchically structured composite of HA mineral reinforced by the collagen fibrils (Kohles and Martinez, 2000). The HA micro crystals are embedded into the collagen fibrils forming the mineralized collagen fiber. There are at least three typical hierarchical levels in cortical bone. They are the collagen-HA crystal composite with micro spaces between the crystallites (20–60 nm); the layered lamellae with lacunae-canaliculi pores (0.1 μm); and the arrangement of lamellae into cylindrical structures (osteons) with Volksmann canals and Haversian lumens (20–50 μm).

The hierarchical structure of bone explains that the electric polarization types occurring in it are more than typical dipolar and ionic polarization, whose polarization time is much shorter than the order of a millisecond (Jonscher, 2008).

The bone meets the Maxwell–Wagner polarization geometrical conditions. Maxwell–Wagner polarization, also known as "interfacial polarization" or "space charge polarization", occurs in a composite composed of segregated constituents with different dielectric permittivity and electrical conductivity at the inner interfaces of different constituents (Wagner, 1914). Besides the geometric porous structure, the electrical conductivities of collagen and HA in bone are in different orders of magnitude (Andrabi and Behari, 1981), it is reasonable to take the interfacial polarization into consideration. Different with the dipolar and ionic polarization, the polarization time of the Maxwell-Wagner polarization amounts to 10^{-1} s to several hours.

Although the viscoelasticity of dry bones is weak, it may also affect the polarization of collagens in a certain extent. When the collagens in bone are subjected to an electric field, the rotation of the electric dipoles or polarization in the collagens may be resisted by the viscoelasticity, which may slow down the polarization speed of bone. Although we cannot determine the affection exactly, it is equivalent to change the electric properties of collagens, which is one of the reasons for the formation of Maxwell–Wagner polarization.

Hence, the relative slow Maxwell-Wagner polarization is responsible for the '*delay increase*'. During the initial phase of the electric field loading, the dielectric polarization of the bone specimen, between the two electrodes, increases relatively slowly, and then the amount of polarization electric charges accumulated on the specimen surfaces increases slowly accordingly, which leading to the electric field attracted force acted on bone specimen by the electric field increases slowly. Therefore, the bending deflection of a bone specimen with collagens increases slowly at the initial phase of electric field loading.

When the collagens are removed from bone, the remaining inorganic constituent is mainly the basic calcium phosphate HA. Hence, the relatively slow Maxwell–Wagner polarization is not the dominant polarization type any more in HA and only relative faster dipolar polarizations occur as it is inside an electric field. This is the reason why the bending deflection of a burned bone specimen increases rapidly to its peak value once the electric field is applied.

4.2.2. Complex actions of electromechanical properties in bone

The last issue is to comprehend the reason of '*relaxation*'. Why the bending deflection of bone, with and without collagens, cannot maintain constant in constant electric field. To determine that is the general behavior of dielectrics or the inherent property of the bone, comparative experiments were carried out. We measured the deflection response of two cantilevers made of two typical dielectric materials: the polymethyl methacrylate (PMMA) and the polytetrafluoroethylene (PTFE), in the same conditions as the bone underwent in this study. The results are shown in Fig. 8, and the purple and green curves denote the deflection response of the PMMA and PTFE respectively. The responses of the two materials are similar. Once the voltage is applied, their deflections increase to their constant values almost immediately (without delay), and keep constant in the constant electric field. When the electric field is removed, the deflections drop to zero immediately. Another difference of the responses is that the bending directions of PMMA and PTFE depend on the sign of the electric field. When the voltage applied to the electrodes is switched from positive to negative, the deflections of the two materials also turn to the opposite direction.

This study does not intend to investigate the dependence of bending direction of PMMA and PTFE on the sign of electric field. However, their experimental results demonstrate that the bending deflections are constant in a constant electric field, which means that bone has its own dielectric feature, though we can not determine whether this is bone's particular characteristic now. Next, we try to make it clear on the understanding about the '*relaxation*' in bone.

It has been found that Piezoelectric effect (Dragan, 1998; Fukada and Yasuda, 1957), electrostrictive effect (Watanabe et al., 2003; Xu et al., 2015b) and flexoelectric effect (Maranganti et al., 2006) are the three electromechanical coupling effects in some dielectric materials, which can cause electric polarization in dielectrics. Electrostrictive and flexoelectric effects can be found in any general dielectrics. The flexoelectric effect mainly occurs in nano materials, and in most dielectrics the actions of flexoelectricity are too weak to be tested.

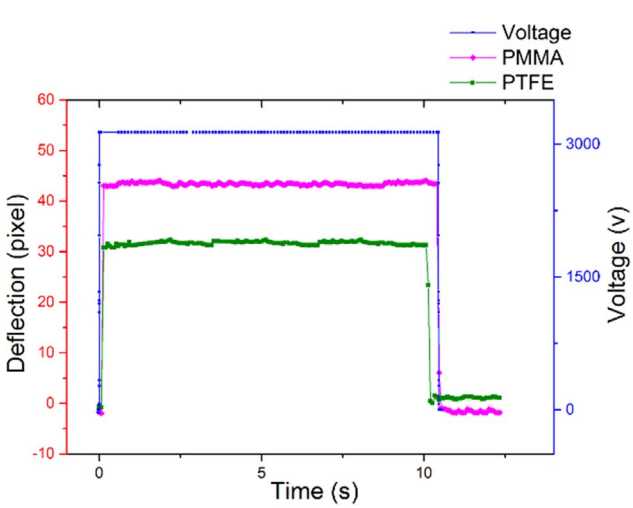

Fig. 8. Deflections of PMMA specimen and PTFE specimen versus time. Blue denotes the waveform of the applied voltage. Purple denotes the deflections of the PMMA specimen and green denotes the PTFE specimen respectively.

As mentioned, the reason causing the bending of bone is the electrostriction effect for which the polarization of bone is proportional to the square of electric field. Literature Xu (Xu et al., 2015b) concludes a formula that the deflection of bone is proportional the square of the voltage applied. The piezoelectricity is also a bone inherent property that is a first order effect, for which the deformation of bone caused by converse piezoelectricity is proportional to the intensity of electric field. Because bone is a kind of nano materials, the flexoelectric effect exists in bone, for which the polarization via flexoelectricity is proportional to the gradient of strains.

The deflection caused by the converse piezoelectricity of bone was not obtained in our experiments. If that were detected, the total deflections measured would be the superposition of the two displacements caused by electrostriction and piezoelectricity respectively, and the deflections induced by positive and negative voltage would not be identical. However, the deflections measured corresponding to the positive and negative voltages are almost equal, which suggests that the piezoelectricity of bone does not contribute the bending deflection or its contribution is too small to be detected by the setup used. It should be noted that if an alternative electric field is applied to the electrodes, the bone cantilever can be caused to vibrate. While the cantilever of bone is in the resonance state, its vibration amplitudes of the bending deflection are must greater than those obtained in the above experiment. However, the digital image acquisition rate of the system used in this study is only 15 fps, which is too slow to capture sufficient images for the analysis of vibration data.

The flexoelectric effect in bone may play a role for the '*relaxation*'. When a bone subjected to an external electric field inducing nonuniform deformation in its bulk, the strain gradient can induce electric polarization via the flexoelectricity effect.

When the bone specimens with and without collagens become bended, induced by an electric field, the strain gradients across the thickness of the specimens (versus x coordinate, Fig. 1) occur and especially the strain concentrations are even higher around the microspaces in bone. The flexoelectric polarization electric charge will accumulate on the bone specimen's surfaces facing the electrodes, for which the charges are additional besides those induced by the electric field based on electrostriction. However, the sign of the charges induced by flexoelectricity is just opposite to the sign of those caused by the electrostriction (Abdollahi et al., 2014; Mosgaard et al., 2014). In other words, the electric charges on the bone surfaces upon by flexoelectricity counteract the charges upon by electrostriction.

When an electric field is applied, the bone specimen just begins to bend and there are no strain gradients in the bulk of bone. After the electric field becomes constant the bent bone specimen would keep bending in constant, if there were no flexoelectricity in the bone. In fact, the electric charges induced by the bending strain gradients in bone decrease the polarization intensity, thus, the electric force on the specimen becomes weak, and then the specimen will spring back to its initial equilibrium position gradually. This may be the real reason for the '*relaxation*'. We can say that the strain gradient in bone can impact the moving status, when the bones with and without collagens are in an electric field.

Considering the time constant τ_2 in Eq. (1) and τ_c in Eq. (2), they are similar in value, and the time constant τ_c represents the '*relaxation*' of burned bone or HA in bone. The time constant τ_2 represents mainly the '*relaxation*' of bone in the portion beyond the maximum point of the bending curve according to Figs. 2 and 3. By analogy, the '*relaxation*' can also be caused by HA in bone.

Then, we can say that the organic collagens decrease the electric polarization rate of bone, which leads to the '*delay increase*' and '*relaxation*' lies on the inorganic component of bone.

The ultimate purpose of the study on electromechanical properties of bone is to understand bone's electric behavior for clinical application. Both dry and wet bones are dielectrics (Kosterich et al., 2007; Reddy and Saha, 1984), and they both have piezoelectricity, electrostriction and other electromechanical properties belong to dielectrics (Minaryjolandan and Yu, 2009; Xu et al., 2015a). It can be say that the electromechanical behaviors of dry bone reflect the electromechanical properties of wet bone to a certain extent.

5. Conclusions

The electromechanical properties of bone manifest in several aspects. When subject to a square waveform electric field, the bending response of a bone cantilever with time is different from that of a bone specimen without collagens. The bending deflection response of a bone cantilever with collagens shows a delay increasing feature, which takes about one second for reaching its maximum deflection, whereas the bending deflection increases almost immediately to its maximum value within the same electric field, when the collagens are removed from the specimen.

An additional experiment demonstrates that the electric polarization induced by an external electric field, is from Maxwell–Wagner polarization mechanism, for which the polarization rate is relatively slow and leads to the delayed increasing deflection.

The second finding is that an electric polarized bending bone cantilever, in a constant electric field, springs back to its initial position gradually instead of holding bending constantly like general dielectrics. Another additional experiment demonstrates that the flexoelectricity effect in bone hydroxyapatite governs the behavior.

Acknowledgment

The work was supported by the National Natural Science Foundation of China under Grant No. 11372218.

References

Abdollahi, A., Peco, C., Millan, D., Arroyo, M., Arias, I., 2014. Computational evaluation of the flexoelectric effect in dielectric solids. J. Appl. Phys. 116, 435.
Anderson, J.C., Eriksson, C., 1968. Electrical properties of wet collagen. Nature 218, 166–168.
Anderson, J.C., Eriksson, C., 1970. Piezoelectric properties of dry and wet bone. Nature 227, 491–492.
Andrabi, W.H., Behari, J., 1981. Formation of bone electrets and their charge decay characteristics. Biomaterials 2, 120–123.
Aschero, G., Gizdulich, P., Mango, F., Romano, S.M., 1996. Converse piezoelectric effect detected in fresh cow femur bone. J. Biomech. 29, 1169–1174.
Atsushi, H., 2015. Numerical simulation of piezoelectric effect of bone under ultrasound irradiation. Jpn. J. Appl. Phys. 54, 07HF06.
Bing, P., Kemao, Q., Huimin, X., Anand, A., 2009. Two-dimensional digital image correlation for in-plane displacement and strain measurement: a review. Meas. Sci. Technol. 20, 062001.
Dragan, D., 1998. Ferroelectric, dielectric and piezoelectric properties of ferroelectric thin films and ceramics. Rep. Progress. Phys. 61, 1267.
Figueiredo, M., Fernando, A., Martins, G., Freitas, J., Judas, F., Figueiredo, H., 2010. Effect of the calcination temperature on the composition and microstructure of hydroxyapatite derived from human and animal bone. Ceram. Int. 36, 2383–2393.
Flügge, D.-I.W., 1975. Viscoelasticity. Springer, Berlin Heidelberg.
Fukada, E., Yasuda, I., 1957. On the piezoelectric effect of bone. J. Phys. Soc. Jpn. 12, 1158–1162.
Halperin, C., Mutchnik, S., Agronin, A., Molotskii, M., Urenski, P., Salai, M., Rosenman, G., 2004. Piezoelectric Effect in Human Bones Studied in Nanometer Scale. Nano Lett. 4, 1253–1256.
Hastings, G.W., Mahmud, F.A., 1988. Electrical effects in bone. J. Biomed. Eng. 10, 7.
Hou, Z., Fu, D., Qin, Q.-H., 2011. An exponential law for stretching–relaxation properties of bone piezovoltages. Int. J. Solids Struct. 48, 603–610.
Isaacson, B.M., Bloebaum, R.D., 2010. Bone bioelectricity: what have we learned in the past 160 years? J. Biomed. Mater. Res. Part A 95A, 1270–1279.
Itoh, S., Nakamura, S., Kobayashi, T., Shinomiya, K., Yamashita, K., 2006. Effect of electrical polarization of hydroxyapatite ceramics on new bone formation. Calcif. Tissue Int 78, 133–142.
Jonscher, A.K., 2008. Dielectric relaxation in solids. Xi'an Jiaotong University Press.
Kohles, S.S., Martinez, D.A., 2000. Elastic and physicochemical relationships within cortical bone. J. Biomed. Mater. Res. Part B Appl. Biomater. 49, 479–488.
Kosterich, J.D., Foster, K.R., Pollack, S.R., 2007. Dielectric permittivity and electrical conductivity of fluid saturated bone. IEEE Trans. Biomed. Eng. BME 30, 81–86.
Maranganti, R., Sharma, N.D., Sharma, P., 2006. Electromechanical coupling in nonpiezoelectric materials due to nanoscale nonlocal size effects: green's function solutions and embedded inclusions. Phys. Rev. B 74, 014110.
Minaryjolandan, M., Yu, M.F., 2009. Uncovering Nanoscale electromechanical heterogeneity in the Subfibrillar structure of collagen fibrils responsible for the piezoelectricity of bone. ACS Nano 3, 1859–1863.
Mosgaard, L.D., Zecchi, K.A., Heimburg, T., 2014. Electrical properties of polar membranes. Physics.
Nashif, A.D., Jones, D.I.G., Henderson, J.P., 2001. Handbook of Viscoelastic Vibration Damping.
Nicholson, R.A., 1992. Bone survival: the effects of sedimentary abrasion and trampling on fresh and cooked bone. Int. J. Osteoarchaeol. 2, 79–90.
Pienkowski, D., Pollack, S.R., 1983. The origin of stress-generated potentials in fluid-saturated bone. J. Orthop. Res. 1, 30–41.
Qin, Q.H., Qu, C.Y., Ye, J.Q., 2005. Thermo electroelastic solutions for surface bone remodeling under axial and transverse loads. Biomaterials 26, 6798–6810.
Qin, Q.H., Ye, J.Q., 2004. Thermoelectroelastic solutions for internal bone remodeling under axial and transverse loads. Int. J. Solids Struct. 41, 2447–2460.
Qu, C.Y., Qin, Q.H., Kang, Y.L., 2006. A hypothetical mechanism of bone remodeling and modeling under electromagnetic loads. Biomaterials 27, 4050–4057.
Raspanti, M., Guizzardi, S., De Pasquale, V., Martini, D., Ruggeri, A., 1994. Ultrastructure of heat-deproteinated compact bone. Biomaterials 15, 433–437.
Reddy, G.N., Saha, S., 1984. Electrical and dielectric properties of wet bone as a function of frequency. IEEE Trans. Biomed. Eng. 31, 296–303.
Ren, L., Yang, P., Wang, Z., Zhang, J., Ding, C., Shang, P., 2015. Biomechanical and biophysical environment of bone from the macroscopic to the pericellular and molecular level. J. Mech. Behav. Biomed. Mater. 50, 104–122.
Ren, C., Hou, Z., Zhao, W., 2009. Stiffness compatibility of coralline hydroxyapatite bone substitute under dynamic loading. Sci. China Ser. E-Technol. Sci. 52, 1331–1336.
Rosa, N., Simoes, R., Magalhães, F.D., Marques, A.T., 2015. From mechanical stimulus to bone formation: a review. Med. Eng. Phys. 37, 719–728.
Sutton, M.A., Cheng, M., Peters, W.H., Chao, Y.J., McNeill, S.R., 1986. Application of an optimized digital correlation method to planar deformation analysis. Image Vision. Comput. 4, 143–150.
Wagner, K., 1914. Erklärung der dielektrischen Nachwirkungsvorgänge auf Grund Maxwellscher Vorstellungen. Archiv f Elektrotechnik 2, 371–387.
Watanabe, M., Wakimoto, N., Shirai, H., Hirai, T., 2003. Bending electrostriction and space-charge distribution in polyurethane films. J. Appl. Phys. 94, 2494–2497.
Wieland, D.C.F., Krywka, C., Mick, E., Willumeit-Römer, R., Bader, R., Kluess, D., 2015. Investigation of the inverse piezoelectric effect of trabecular bone on a micrometer length scale using synchrotron radiation. Acta Biomater. 25, 339–346.
Xu, L., Hou, Z., Fu, D., Qin, Q.-H., Wang, Y., 2015a. Stretched exponential relaxation of piezovoltages in wet bovine bone. J. Mech. Behav. Biomed. Mater. 41, 115–123.
Xu, L., Hou, Z., Fu, D., Yang, L., Yi, W., Kang, H., 2015b. Measurement of electrostriction in bone using digital image correlation. Opt. Lasers Eng. 65, 3–8.