Program Digest, 2019 IEEE International Ultrasonics Symposium (IUS)
Glasgow, Scotland, October 6-9, 2019

A simple model for the simulation of ultrasonically induced electric potentials

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***Abstract*— Although ultrasonically induced electric potentials in bone may play an important role in the bone formation and remodeling process, many aspects of the bone's weak piezoelectricity are still unclear. Most of the previously conducted studies consider a biological rather than electrical perspective. In this study, the ultrasonically induced electric potentials were experimentally measured by the bone transducer, and then compared to those numerically derived. To better understand the mechanoelectrical response of bone, a simple model for the simulation of ultrasonically induced electric potentials is here proposed. This model allows a quantitative analysis of the ultrasonic effects on bone, to investigate some of its interesting properties, such as inhomogeneity. The equivalent circuit parameters were obtained by linear and least squares method fitting, and proved to give a reasonably accurate, effortless simulation.**

Keywords*— *piezoelectricity, ultrasonic transducer, bone, equivalent circuit

I. Introduction

In 1953, Fukada and Yasuda reported that electrical potentials were induced in the bone, when subject to a mechanical stress. Additionally, Fukada and Yasuda proved that the stress-induced electrical potentials are due to the bone's piezoelectric properties [1]. In another study, Shamos and Lavine provided evidence that the bone induced electrical potentials promote bone growth and speed up the healing process of a fracture [2]. Because of the important role in the bone formation and remodeling process that these electric potentials concern, the piezoelectric behavior of bone should be clearly understood. However, the biological response of bone to ultrasound irradiation, as reported by Padilla et al., is complex as numerous cell types respond to this stimulus involving several pathways [3]. Moreover, due to the complex shape, anisotropy, and heterogeneity of bone, the effects of a mechanical stress on the bone are difficult to evaluate. Although several mechano-biological studies on the mechanical sensing systems of cells, membranes and proteins have been conducted, most of them concerned low frequency mechanical stresses [4],[5]. One exception is the study by Okino et al. [6], which demonstrates that bone can generate electric potentials by ultrasound irradiation in the MHz range, at much higher frequencies than those considered in the previous studies. Ikushima also reported radiation of electromagnetic wave from bone due to the ultrasonic radiation [7]. At higher frequencies, as the study by Tsuneda et al. confirms [8], the adoption of a new electrical perspective in the investigation of the bone's piezoelectricity could prove beneficial. To evaluate the bone's weak piezoelectricity, one possible solution is the bone ultrasound receiver. To better understand the mechanoelectrical response of bone, a simple equivalent circuit of the receiver was proposed [9]. Moreover, interesting mechanical and electrical properties of the bone, such as inhomogeneity, or the piezoelectric constants, could be discovered by a quantitative analysis of the simulated potentials.

II. Fabrication And Setup

To evaluate the stress induced electric potentials, ultrasonic transducers using bovine cortical bone as its piezoelectric material were fabricated [6],[8]. Two transducers, namely A and B, were constructed. The ultrasonically induced potentials detected as the output of the fabricated transducers were almost 1/1000 of that of a conventional polyvinylidene (PVDF) transducer [10]. In this study, we also fabricated the transducer and evaluated its characteristics. From the mid shafts of bovine femurs, disk samples were obtained. These disks are 10.0-10.5 mm in diameter and 1.00±0.01 mm thick. The fabrication process of the bone sample is shown in Figure 1. For the receiver, the bone disks were pasted onto the surface of a brass cylinder (backing material). Next, gold was deposited on the bone sample to make the electrode. For the transmitter a transducer with a PVDF film was also fabricated [10].

In the ultrasonic experiments, the transmitter and the receiver were set coaxially in a water tank at the room temperature. At the time of the measurements, the bone samples had been in water for more than 30 minutes. A function generator (33250A; Agilent Technologies, Santa Clara, California) generated a 10-cycle sinusoidal burst wave in the frequency range from 800 kHz to 2.4 MHz. The mechanical resonance of the receiver was obtained from the ratio of the longitudinal wave velocity and its wavelength. Because the thickness of the sample corresponds to half of the wavelength, the wavelength is equivalent to twice the sample's thickness. The speed of sound in bone was approximately 4000 m/s [11], and the thickness of the sample was 1 mm, so the first resonance frequency was around 2 MHz. The input signal's frequencies considered are therefore below, at and above the resonance. The input burst wave was then amplified to 70 Vpp by a bipolar power supply (HAS 4101; NF, Yokohama, Japan). Finally, the received signals were further amplified to 40 dB by a pre-amplifier (BX-31A; NF, Yokohama, Japan) and observed by an oscilloscope (DPO3054; Tektronix, Beaverton, Oregon). This experimental setup is shown in Figure 2. The transmitted ultrasound pressure at the measurement point was around $10\ \mathrm{kPa}_{\mathrm{peak-peak}}$.

To simulate the experimentally measured waveforms, the simple model in Figure 3 was used. This model – known as Van Dyke Model - is the most basic model characterizing a piezoelectric transducer near the resonance frequency. It is typically used to model electromechanical resonance characteristics of crystal oscillators [12]. It consists of: E_0, which represents a voltage proportional to the applied pressure, L, mechanical mass, R, damping, C, compliance, and C_0, that represents the electrical capacitance.
To derive the circuit's parameters two different fitting methods were employed: linear and least squares.

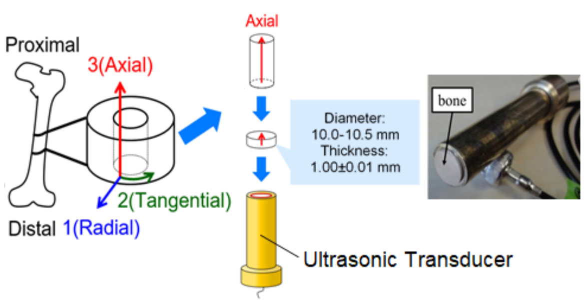


Figure 1 Fabrication process of bone disk samples and bone transducers.

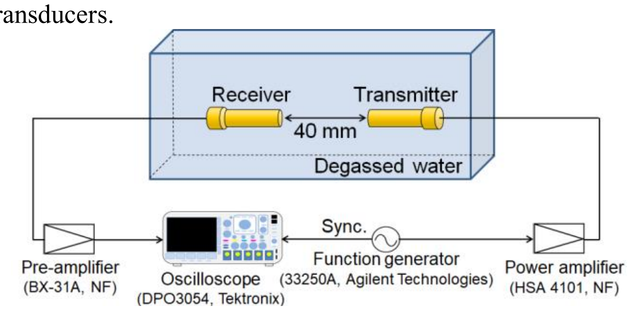


Figure 2 Experimental Setup: the transmitter is a PVDF transducer, the receiver is a bone transducer. The input signal to the transmitter is a sinusoidal burst wave, 7 Vpp, 10 cycles, 800 kHz – 2.4 MHz.

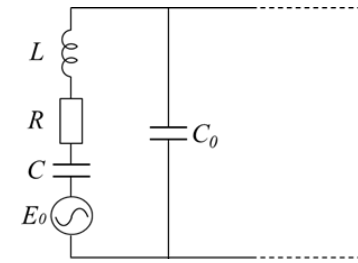


Figure 3 A simple equivalent circuit model of the bone transducer.

III. Numerical Simulations for the Parameters

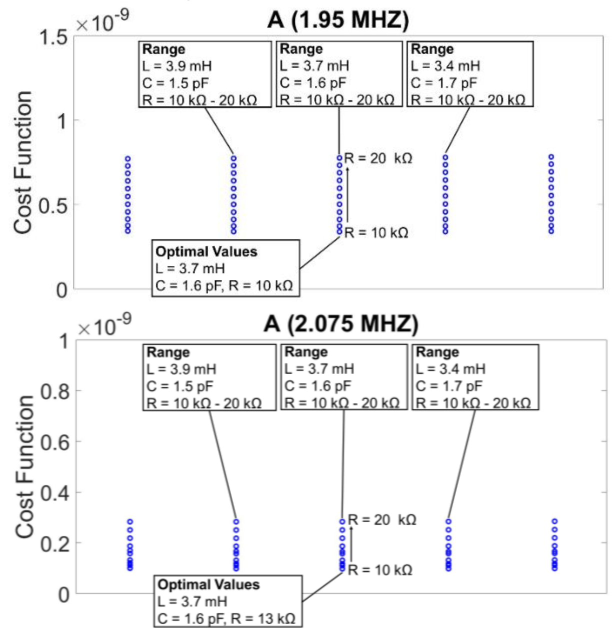
To obtain C_0, the admittance curve of the ultrasound bone transducer was measured by an impedance analyzer (E4990, Keysight).
From the s-domain description of the model's admittance

$$Y = sC_0 + \frac{1}{sL+\frac{1}{sC}+R} \qquad (1)$$

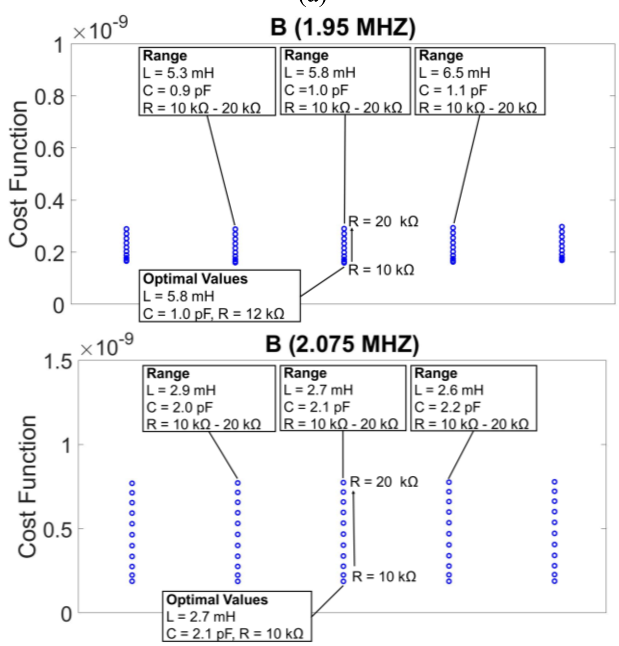
a reasonable estimation for C_0 can be found. Because the resonance peak of the admittance is invisible, Q is estimated to be very small (< 0.1), and therefore only the first term can be considered [13]. The linear fitting yielded C_0 of around 13.3 pF. The values of L, C and R were instead obtained by least squares fitting method of the simulated potentials to those experimentally detected, as suggested by Terunuma and Nishigaki in their study [14]. The idea behind this method is that the sum of the squares of the difference between the measured and simulated voltages can be minimized by sweeping L, C, and R over a range of values and finding the optimal ones. The equation to be minimized is referred to as cost function D:

$$D = \sum_{i=1}^{N}(v_{sim,i} - v_{meas,i})^2 \qquad (2)$$

where N is the number of samples, and v_{sim}, v_{meas} are the simulated and detected potentials, respectively. The optimal values are instead the L, C and R values for which the cost function is minimized. Figure 4 shows two examples of cost functions: at every new iteration the value of C is increased by 0.1 pF, while within each iteration the value of C is kept unchanged and the value of R is increased with steps of 1 kΩ. This explains the repetitive, broken trend of the cost functions in Figure 4. For a circuit to accurately simulate piezoelectric resonance behavior, the value of C must be smaller than C_0. Therefore, for C, values in the range 10 fF to 20 pF were tested. The corresponding values of L were given by the relation $f_{res} = \frac{1}{2\pi\sqrt{LC}}$, where f_{res} is the resonance frequency of the bone transducer (around 2 MHz). For R, instead, values within 10 kΩ to 20 kΩ were used. This choice is partly motivated by a study on the factors affecting the measurement of bone impedance conducted by Saha et al., which suggested that the average specific resistance of a bone specimen is around 20 kΩcm [15]. Given a 1mm thick sample, according to the average specific resistance, R should be ~ 2 kΩ. However, many factors were found to greatly affect the resistance value (up to 400% changes). In our case, much better results were obtained with higher R values in the order of 10 kΩ. To understand the quality of this fitting, comparisons between measured and simulated waveforms - when these optimal values are plugged into the equivalent model - were plotted. Four examples of this comparisons are represented in Fig. 5. Experiments and numerical results are in good agreement. Figure 5 shows the waveforms at 1.95 MHz and 2.075 MHz, which are below and at the resonance frequency, respectively. Unfortunately, the detected waveforms at frequencies well below or above resonance were very small and buried in noise and, therefore, unsuitable for fitting.

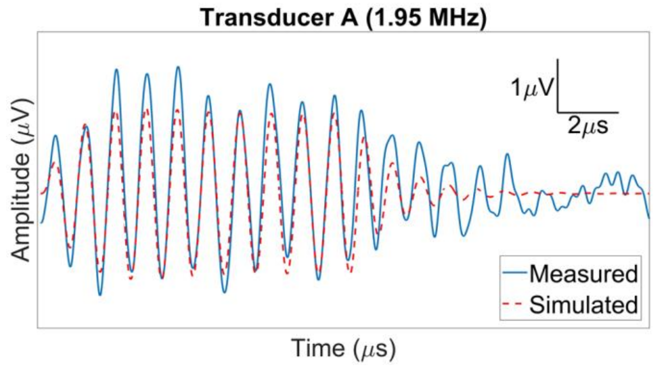


(a)

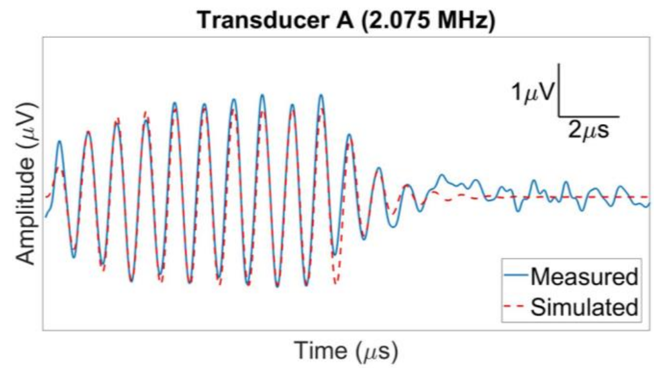


(b)

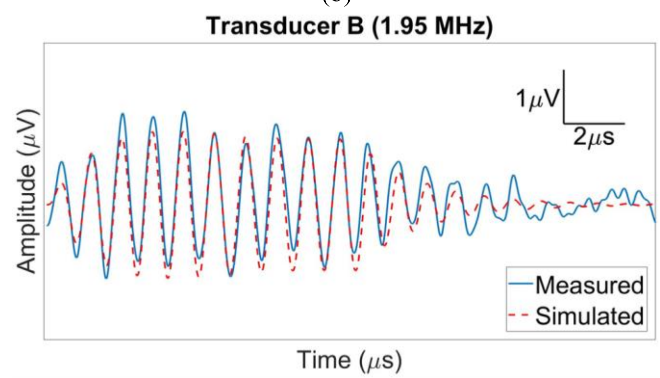
Figure 4 (a) cost function of transducer A at 1.95 MHZ and 2.075 MHz (b) cost function of transducer B at 1.95 MHz and at 2.075 MHz.



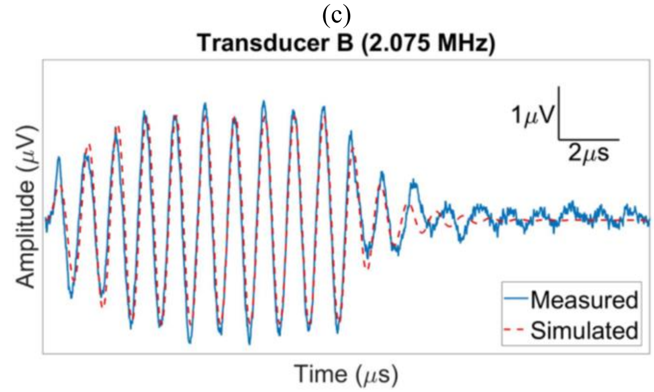
(a)



(b)



(c)



(d)

Figure 5 (a), (b) Comparison of measured and simulated waveforms for transducer A at 1.95 MHz (below the resonance frequency) and 2.075 MHz (at the resonance frequency). (c), (d) Comparison of measured and simulated waveforms for transducer B at 1.95 MHz and 2.075 MHz.

IV. DISCUSSIONS

By linear and least squares fitting of the simulated waveforms to those experimentally measured, the circuit's component's optimal values were derived. When plugging these optimal values in the model, simulations and experiments are in good agreement (Fig. 5 (a), (b), (c) and (d)), indicating that the derived parameters will give a reasonably accurate, effortless simulation to better understand the mechanoelectrical response of bone. The derived values proved to be suitable for all waveforms - around the resonance - measured by the same transducer, although with slightly different tolerances. For transducer A, the optimal values of C and L were precisely the same at all frequencies, whereas for transducer B the derived values did differ slightly (from Figure 4 (b), C = 1.0 pF and C = 2.1 pF). This difference is probably due to the simplicity of the lumped parameter model that was used in the simulations. At high frequencies, the irradiated wavelength becomes much smaller than the sample's thickness and, to obtain a better accuracy, a distributed model (like the Mason model [16]) should be considered. R was determined with a tolerance of $\pm$ 3 kΩ for both transducer A and transducer B. These values are reported in Table 1. It is interesting to note that, the closer the frequency is to resonance, the more accurate the model is. A slight discrepancy between the tails of the simulated and measured waveforms is noticeable and it may be due to a difference in the actual and derived Q factor values. Moreover, an interesting contradiction is highlighted, as the Q factor of the equivalent circuit model (around 7) was much higher than that expected from the bone's admittance curve. This discrepancy is possibly related to the bone's heterogeneity [11]. The water content in the bone may also have a relevant role in determining the equivalent R of the circuit (thus the Q factor). As an example, at low frequency (20 Hz), a value of around 66 MΩ was measured when the sample was dry, whereas in wet conditions that value dropped to around 64 kΩ.

Table 1. Optimal values for transducers A and B at 1.95 MHz and 2.075 MHz.

Transducer	Freq. (MHz)	L (mH)	C (pF)	R (kΩ)	C_0 (pF)
A	1.95	3.67	1.6	10	13.3
A	2.075	3.67	1.6	13	13.3
B	1.95	5.8	1.0	12	13.3
B	2.075	2.7	2.1	10	13.3

V. CONCLUSION

The piezoelectric character of bone was experimentally discussed. The simple model showed good agreements with the character, revealing the possibility that bone's properties may be evaluated from an electric point of view.

REFERENCES

[1] A. Fukada, et al., J. Phys. Soc. Jpn., **12**, 1158 (1957).
[2] M.H. Shamos, L.S. Lavine, Clin. Orthop, **35**, 177 (1964).
[3] F. Padilla et al., Ultrasonics, **54**, 1125 (2014).
[4] I. Iscla, P. Blount, Biophys. J., **103**, 169 (2012).
[5] S.S. Ranade, et al., Neuron, **87**, 1169 (2015).
[6] M. Okino, et al., Appl. Phys. Lett., **103**, 103701 (2013).
[7] K. Ikushima, et al., Appl. Phys. Lett., **89**, 194103 (2006).
[8] H. Tsuneda, et al., Appl. Phys. Lett., **106**, 073704 (2015).
[9] S. Sherrit et al., Proc. IEEE Ultrason. Symp., 931 (1997).
[10] Y. Nakamura, T. Otani, J. Acoust. Soc. Am., **94**, 1191 (1993).
[11] Y. Yamato, et al., IEEE, Trans. Ultrason. Ferroelect. Freq. Cont., **55**, 1298 (2008).
[12] J. Kina et al., Proc. IEEE Sensors Appl. Symp., (2008).
[13] T. Makino, et al., IEEE, Trans. Ultrason. Ferroelect. Freq. Cont. Submitted.
[14] K. Terunuma, S. Nishigaki Jpn. J. Appl. Phys., Supple. **22-2**, 143 (1983).
[15] S. Saha et al., Med. Biol. Eng. Comput., **22**, 123, (1984).
[16] J.L. San Emerterio, et al., *Piezoelectric transducers and applications*, Chap. 3, Springer-Verlag, 2004.