Slow and fast ultrasonic wave detection improvement in human trabecular bones using Golay code modulation

Bahman Lashkari[a)]
Center for Advanced Diffusion-Wave Technologies (CADIFT), Department of Mechanical and Industrial Engineering, University of Toronto, Toronto, Ontario, Canada
bahman@mie.utoronto.ca

Amir Manbachi
Institute of Biomaterials and Biomedical Engineering (IBBME), University of Toronto, Toronto, Ontario, Canada
amir.manbachi@utoronto.ca

Andreas Mandelis
Center for Advanced Diffusion-Wave Technologies (CADIFT), Department of Mechanical and Industrial Engineering, University of Toronto, Toronto, Ontario, Canada
mandelis@mie.utoronto.ca

Richard S. C. Cobbold
Institute of Biomaterials and Biomedical Engineering (IBBME), University of Toronto, Toronto, Ontario, Canada
cobbold@ecf.utoronto.ca

Abstract: The identification of fast and slow waves propagating through trabecular bone is a challenging task due to temporal wave overlap combined with the high attenuation of the fast wave in the presence of noise. However, it can provide valuable information about bone integrity and become a means for monitoring osteoporosis. The objective of this work is to apply different coded excitation methods for this purpose. The results for single-sine cycle pulse, Golay code, and chirp excitations are compared. It is shown that Golay code is superior to the other techniques due to its signal enhancement while exhibiting excellent resolution without the ambiguity of sidelobes.

PACS numbers: 43.80.Ev, 43.80.Vj, 43.80.Jz [CC]
Date Received: June 21, 2012 **Date Accepted:** July 23, 2012

1. Introduction

Bone ultrasound has found extensive and growing applications in diagnostic medicine in the past few decades.[1] However, the current state-of-the-art demands further theoretical and experimental developments. Cancellous bone has a very complex structure consisting of a matrix of connected plates and rods, called trabeculae. These spongy structures are interspersed with marrow. The trabeculae are not arranged uniformly, but tend to align in accordance with the stress distribution in the bone. This inhomogeneous, anisotropic composition makes it very difficult to predict and interpret the propagation of acoustic waves in bones. Biot's theory of elastic wave propagation through fluid saturated porous media was the first theory applied to bone ultrasound.[2] Many modifications have been proposed for Biot's theory to increase its prediction accuracy for speed of sound and attenuation.[3,4] Biot's theory, however, does not account for

[a)]Author to whom correspondence should be addressed.

multiple scattering in the bone and its success is restricted to lower frequencies where the acoustic wavelength is larger than the pores. One interesting prediction of Biot's theory is the existence of one transverse and two longitudinal waves. The so-called fast and slow longitudinal waves are assumed to be generated as a result of constructive and destructive interferences between relatively faster and slower waves passing through the solid and liquid medium, respectively.[3,4] The first attempts to apply Biot's theory to the ultrasonic wave propagation in cancellous bone were focused on the phase velocity and attenuation of the slow wave.[5,6] The first experimental observation of two-wave phenomenon in bone was conducted by Hosokawa and Otani on bovine cancellous bone.[7] In that study, the authors used a wideband (0.1–10 MHz) polyvinylidene difluoride (PVDF) transducer and a hydrophone to transmit and detect the ultrasonic wave, and successfully demonstrated a clear correlation between the amplitudes of the two above-mentioned waves and the bone volume fraction. Similar experiments employing two PVDF transducers were performed later on 35 cubic human cancellous bones (extracted from femoral head) and the two-wave phenomenon was observed on 16 specimens.[8] The wave separation occurred in the principal bone stress direction and in two cases it was observable in another direction as well. It was concluded that wave separation strongly depends on the anisotropy of the bone structure as well as the bone volume fraction.[8,9] Recently, the observation of the two-wave phenomenon in a cancellous bone covered by two cortical layers has also been reported.[10] A more comprehensive account of the two-wave observations is summarized in Chap. 11 of Ref. 1.

Although the properties of fast and slow waves can provide valuable information about the integrity of bone structure, few techniques have been proposed to improve the detection of the abovementioned two waves.[11–13] Therefore, further development is needed to set the stage for reliable clinical applications. In this work, we propose the use of Golay codes (GC) as well as other coded excitation methods to increase the ultrasonic signal-to-noise ratio (SNR) while preserving resolution. Practical obstacles in separating fast and slow waves are the weak amplitude of the fast moving wave as well as their possible temporal overlap. Here, we aim to demonstrate how the use of GC can enhance the separation of both waves while still ensuring that the small amplitude fast wave is clearly visible. It should be mentioned that GCs have been used previously in bone ultrasound, however, they have been limited to measurements of the acoustic attenuation.[14]

2. Materials and methods

2.1 GCs in ultrasonics

The coded excitation method has been devised mainly in radar technology in the quest for increasing the signal while employing limited transmission power. This is because pulse compression of the coded waveform enables an increase in the SNR while preserving axial resolution. Many different coded waveforms have been devised. The most efficient ones are frequency modulated chirps and Golay sequences.[15] The former generates the highest SNR while the latter has the advantage of being sidelobe free,[16] though it does require two transmission cycles.

In order to control the frequency content of the GC signal, the binary codes can be implemented by sinusoidal half-cycles [Fig. 1(a)]. However, as shown by Nowicki *et al.*,[17] the bandpass effect of the ultrasonic transducer will be an important factor reducing signal enhancement and axial resolution. To solve this problem, they suggested the use of a full-cycle for each bit as shown in Fig. 1(b).[17] This concept has been shown here through a number of simulations that are illustrated in Fig. 1 for future comparison with experiments. In Figs. 1(a) and 1(b), two GC signals with identical lengths (16-bit half-cycle vs 8-bit full-cycle) are shown. The spectra of these signals are completely different and lead to completely different cross-correlations [Figs. 1(c) and 1(d)]. The cross-correlated signals are compared for three cases of no bandpass filtering as well as 100% and 50% filtering bandwidths [Figs. 1(d)–1(f)]. It is

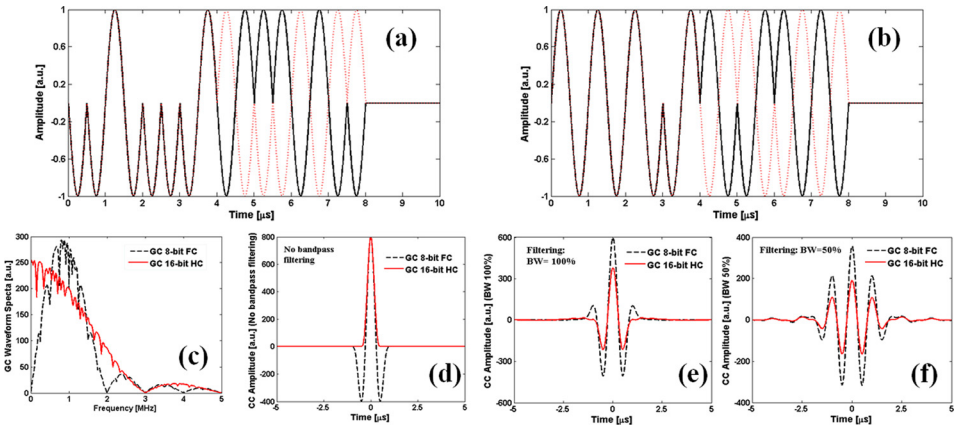

Fig. 1. (Color online) Simulations show the effect of bandpass filtering on signal enhancement and axial resolution of applied GCs. (a) A 16-bit GC employing half-cycle (HC) sine wave and its complementary code, (b) an 8-bit GC employing full-cycle (FC) sine wave and its complementary code. (c) Spectrum of GC signals in (a) and (b), (d) GC cross-correlation from (a) and (b) sequences without any bandpass filtering, (e) filtering with bandwidth (BW) = 100%, and (f) filtering with BW = 50%.

shown that although the full-cycle generates better SNR enhancement, it sacrifices axial resolution [Figs. 1(d) and 1(e)]. It has also been mentioned[17] that in the case of 90% or lower bandpass filtering, the difference between axial resolution of half-cycle and full-cycle GC is insignificant. In our present quest to more clearly identify fast and slow waves that may be partially overlapping, axial resolution is more crucial than SNR.

In our experiments we used 512 bit complementary GCs and the results of both half-cycle bit and full-cycle bit GCs were compared. To generate long GCs, we employed the algorithm described in Appendix A of Ref. 17. In addition, we also report on the application of linear frequency-modulated (LFM) chirp excitation and single-cycle sine pulse. In chirp excitation, it has been previously shown that the ultrasonic transducer acts as filter and reduces the sidelobes.[16,18] In addition, we implemented 5% waveform tapering in order to eliminate spectral ripples and reduce sidelobes even further.[18]

2.2 Experimental setup

The setup for bone ultrasound experiments is shown in Fig. 2. Two ceramic transducers were immersed in water facing each other and the sample was located between them. The transmitter was a focused 1-MHz transducer with focal distance 1.9″ (V314 Olympus Panametrics NDT, Waltham, MA) and the receiver was a flat 1-MHz transducer (A303A Olympus Panametrics NDT). The transmitted signal was amplified using a RF power amplifier (411LA ENI Co., Rochester, NY, 10 W and 40 dB). In some experiments we used a preamplifier (5676 Olympus Panametrics NDT, 40 dB)

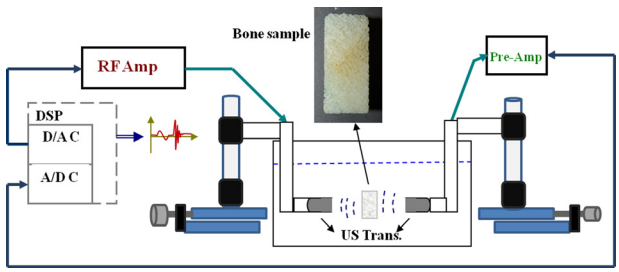

Fig. 2. (Color online) Experimental setup of the bone ultrasonic pitch-catch system.

after the receiver transducer, as discussed in the section on experimental results. Data acquisition and signal processing were performed in the PC using LabView software. To transmit the detected signals to the software, analog-to-digital (ADC, PXIe-5122) and NI-SCOPE software (National Instruments, Austin, TX) were used. Waveform generation was performed by a digital-to-analog converter (DAC, NI PXI-5421) and NI-FGEN software (National Instruments, Austin, TX). The two cards were synchronized using the internal clock of the instrument, therefore the zero delay time in the figures corresponds to the start of the transmitted signal. The maximum sampling frequency of both cards was 100 MHz, which was used in all reported experiments. Experiments were performed on two samples: (1) a glass layered sample and (2) a human calcaneus bone. The glass layered model was constructed from borosilicate microscope slips (Fisher Scientific) with 0.2 mm thickness and $22 \times 22\,\text{mm}^2$ size. The slips were arranged in a staggered pattern resulting in 50% porosity. The human calcaneus bone sample (SteriGraft, San Antonio, TX) was $15.7 \times 15.7 \times 35\,\text{mm}^3$ in size and has an apparent density of $360\,\text{mg/cm}^3$. It was possible to perform ultrasonic tests from two directions along the 15.7 mm length. The temperature of water used as the ultrasonic coupling fluid was $23 \pm 2\,^{\circ}\text{C}$ in the experiments.

3. Experimental results and discussion

The responses to various signal waveforms from glass layered sample are shown in Fig. 3. No preamplifier was used with the receiving transducer. The full-cycle GC generates extra correlation noise which could be mistaken for a fast moving wave [Fig. 3(c)]. Figure 3(d) shows the normalized envelope signals of the results in Figs. 3(a)–3(c) and a LFM chirp with 0.6 to 1.4 MHz bandwidth. This figure clearly shows that when the noise is insignificant and there is no frequency-dependent attenuation, all coded excitation methods enhance the signal amplitude while maintaining similar signal shapes. Of course, this is not the case for *in vivo* or *in vitro* bone ultrasound

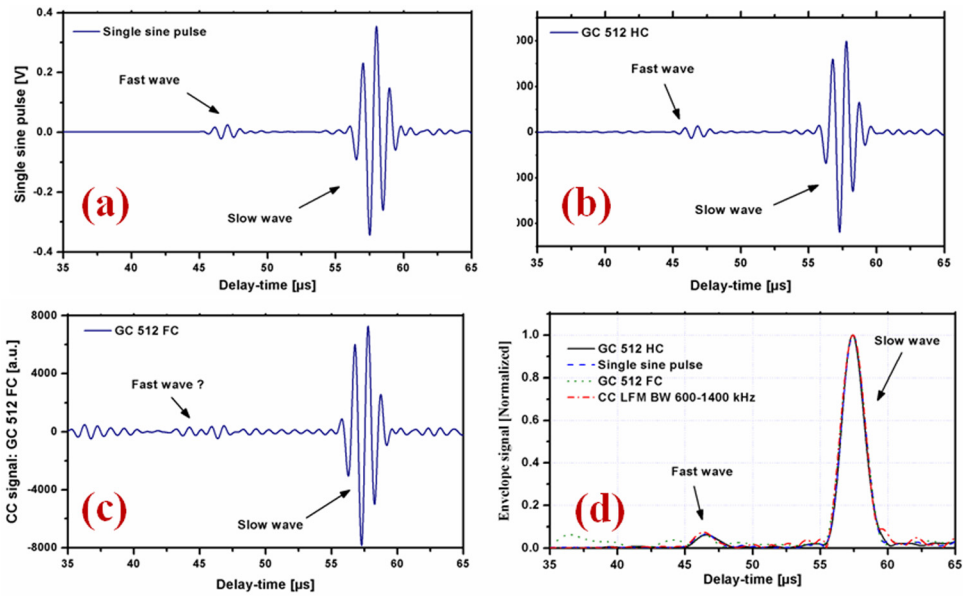

Fig. 3. (Color online) Fast and slow waves detected on a layered glass sample employing (a) single-sine pulse, (b) a 512-bit half-cycle GC, and (c) a 512-bit full-cycle GC. (d) Normalized envelope signal trace of (a) to (c) as well as a LFM chirp.

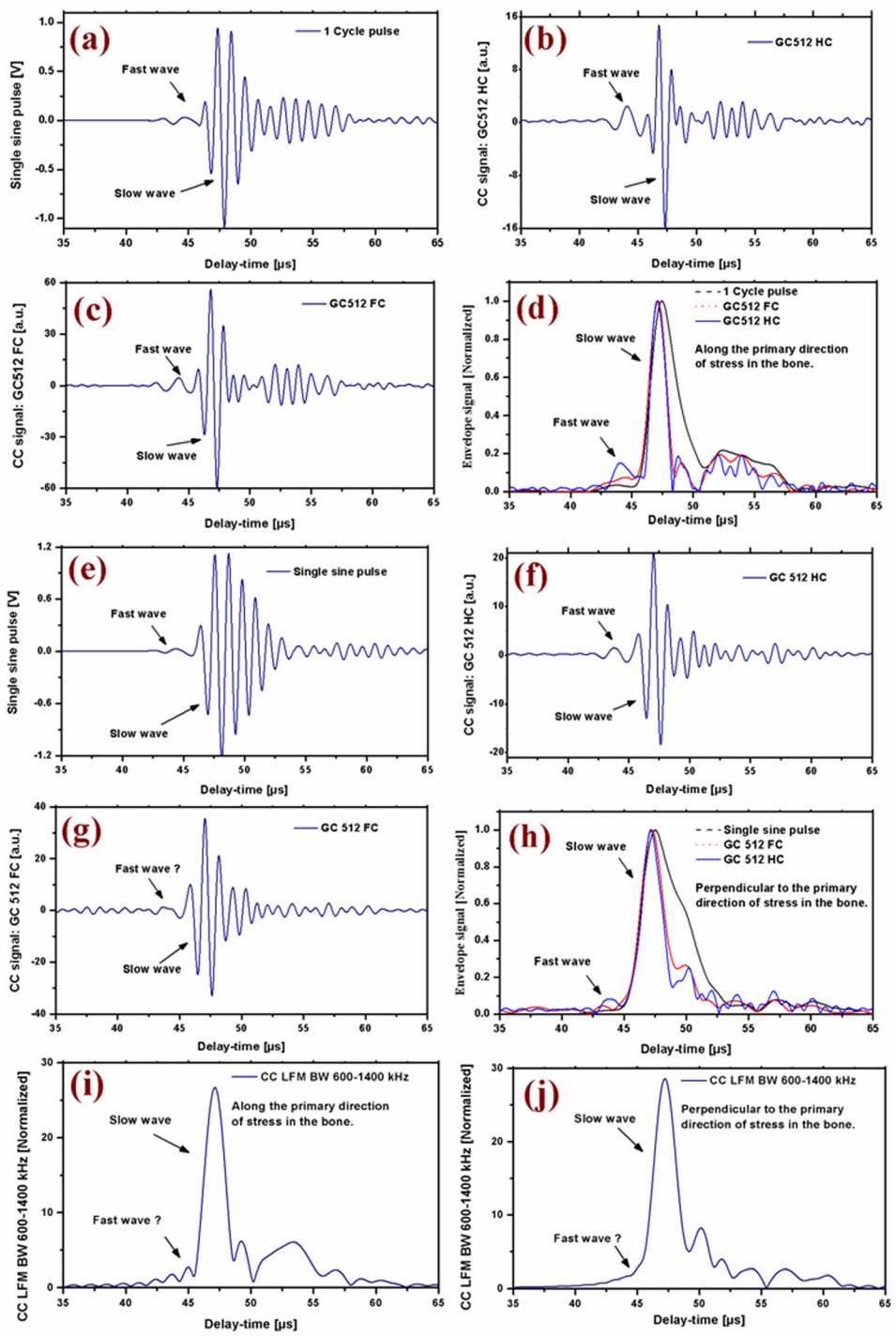

Fig. 4. (Color online) Fast and slow waves detected along the primary stress direction of a human cancellous bone employing (a) single-sine pulse, (b) a 512-bit half-cycle GC, and (c) a 512-bit full-cycle GC. (d) Normalized envelope signal traces from (a) to (c). Signal detection along the direction perpendicular to the primary stress direction employing (e) single-sine pulse, (f) a 512-bit half-cycle GC, and (g) a 512-bit full-cycle GC. (h) Normalized envelope signal traces from (e) to (g). Cross-correlation signals generated by LFM excitation in (i) the primary stress direction and (j) the direction perpendicular to primary stress.

experiments where many noise sources exist. Comparing with signals detected in water with no sample between the transducers, the speed of the slow wave is the same as the speed of sound in water which is consistent with the theory of Johnson *et al.*[19] For a stiff solid the accuracy of approximation $V_{slow}=V_f/\sqrt{\alpha}$ is rather high. Here, V_{slow} is the phase velocity of the slow moving wave, V_f is the speed of sound in the fluid medium, and α is the tortuosity defined by the length of the curve to the distance between its ends (or the average relative length of flow through one side of porous medium to the other[20]), which is unity for a plane. Using the envelope peaks and the length of the sample, the speeds of the fast and slow waves are found to be 3526 and 1491 m/s, respectively. Note that the fast wave speed is measured to be 3526 m/s and 3551 m/s from the half-cycle GC and single-sine signals, respectively.

The responses of the three waveforms associated with the primary direction of stress in the human bone sample are shown in Figs. 4(a)–4(c). A preamplifier was used with the receiving transducer only in the single-sine pulse, Fig. 4(a), because, otherwise, the received signal was too weak for detecting the fast wave. Figure 4(d) shows the normalized envelope signals of the results in Figs. 4(a)–4(c). It can be seen that employing a preamplifier in the single-sine case deteriorates the axial resolution and compromises the ability for wave separation. Using the preamplifier in the case of GC would also compromise the resolution advantage of the coded excitation, however, due to the signal enhancement produced by the pulse compression technique there was no need for extra amplification.

The calculated slow- and fast-wave velocities are 1473 m/s and 2269 m/s, respectively, for single-sine wave excitation, and 1477 m/s and 2058 m/s, respectively, for the half-cycle GC with 512 bits. Unlike the glass model experiment, here the amplitude ratio of the fast to the slow waves and the speed of the fast wave are not the same for single-pulse and GC. This can be attributed to frequency dependent attenuation in the cancellous bone. It is known that the fast wave contains more low frequency content than the slow wave. Therefore, referring to Fig. 1(c), the half-cycle GC with the higher low-frequency content can better resolve the fast wave.

The abovementioned experiment was performed in the perpendicular direction to the primary stress axis and fast and slow waves could also be detected, although not as clearly as in the previous case. The results for single-pulse, half-cycle, and full-cycle 512-bit GC excitation are shown in Figs. 4(e)–4(g). The normalized enveloped signals are also presented in Fig. 4(h). In addition to 1-MHz carrier frequency for GCs, the experiments were repeated at 0.9 and 1.1 MHz. The fast and slow signals were not affected by this minor change. Figures 4(i) and 4(j) show the cross-correlation signal generated by the LFM chirp with frequency range from 600 kHz to 1.4 MHz in the two above mentioned directions. In the measurement along the primary stress direction, Fig. 4(i), some signals other than the main peak are evident which could be hypothesized to be associated with the fast moving wave, although, this would be difficult to confirm. It should be noted that the sidelobes have been reduced by the band-pass filtering effect of the transducer as well as by partially tapering the transmitted signal.[18] In the perpendicular direction to the primary stress, Fig. 4(j), the detection of the fast wave is nearly impossible. Since, the fast wave is completely hidden by the envelope of the cross-correlation signal.

4. Conclusions

In this work the use of various coded ultrasound excitation methods for separating and detecting fast and slow waves in a cancellous bone was investigated. The results demonstrated the ability of Golay codes to be used as a wave identification technique by enhancing the US signal without compromising the axial resolution or generating sidelobes. The spectral content of the coded excitation was found to affect the amplitude ratio of the fast and slow waves. The enrichment of the low frequency content without changing the resolution can help improve the detection of the fast moving wave. It was thus found that the GC with half-cycle is more beneficial in detecting the two-wave phenomenon. In addition, comparing the experiments with the layered glass

structure as a coarse model of a porous medium and with the human trabecular bone showed the importance of the acoustic attenuation factor in contributing to the difficulty to observe the fast and slow waves. Waveform comparisons showed that, although chirp excitation can increase the SNR, sidelobes and the response envelope may hide the small fast moving wave. In conclusion, this work aimed to show that GC can be used as an effective wave identification technique with an ability to improve the possibility of distinguishing fast and slow moving ultrasound waves originating within cancellous bones. This in turn, may be used for better evaluation of bone health, particularly in the diagnosis of osteoporosis.

Acknowledgments

The authors wish to thank Dr. John E. Davies from the University of Toronto (IBBME and Faculty of Dentistry) for providing us with the human bone sample. A.M. (IBBME) acknowledges scholarship support from OGS for his doctoral studies. A.M. (CADIFT) and B.L. gratefully acknowledge the support of the Canada Council for the Arts through a Killam Fellowship to A.M. R.S.C.C. acknowledges NSERC grant support through Grant No. 3297-2012.

References and links

[1] *Bone Quantitative Ultrasound*, edited by P. Laugier and G. H. Haïat (Springer, Dordrecht, 2011).
[2] M. A. Biot, "Theory of propagation of elastic waves in fluid-saturated porous solid. I. Low-frequency range," J. Acoust. Soc. Am. **28**, 168–191 (1956).
[3] T. J. Haire and C. M. Langton, "Biot theory: A review of its application to ultrasound propagation through cancellous bone," Bone **24**, 291–295 (1999).
[4] H. Steeb, "Ultrasound propagation in cancellous bone," Arch. Appl. Mech. **80**, 489–502 (2010).
[5] M. L. McKelvie and S. B. Palmer, "The interaction of ultrasound with cancellous bone," Phys. Med. Biol. **36**, 1331–1340 (1991).
[6] J. Williams, "Ultrasonic wave propagation in cancellous and cortical bone: Prediction of some experimental results by Biot's theory," J. Acoust. Soc. Am. **91**, 1106–1112 (1992).
[7] A. Hosokawa and T. Otani, "Ultrasonic wave propagation in bovine cancellous bone," J. Acoust. Soc. Am. **101**, 558–562 (1997).
[8] K. Mizuno, M. Matsukawa, T. Otani, P. Laugier, and F. Padilla, "Propagation of two longitudinal waves in human cancellous bone: An *in vitro* study," J. Acoust. Soc. Am. **125**, 3460–3466 (2009).
[9] K. Mizuno, H. Somiya, T. Kubo, M. Matsukawa, T. Otani, and T. Tsujimoto, "Influence of cancellous bone microstructure on two ultrasonic wave propagation in bovine femur," J. Acoust. Soc. Am. **128**, 3181–3189 (2010).
[10] K. Mizuno, Y. Nagatani, K. Yamashita, and M. Matsukawa, "Propagation of two longitudinal waves in a cancellous bone with the closed pore boundary," J. Acoust. Soc. Am. **130**, EL122–EL127 (2011).
[11] L. Cardoso, F. Teboul, L. Sedel, C. Oddou, and A. Meunier, "*In vitro* acoustic waves propagation in human and bovine cancellous bone," J. Bone. Miner. Res. **18**, 1803–1812 (2003).
[12] K. R. Marutyan, G. L. Bretthorst, and J. G. Miller, "Bayesian estimation of the underlying bone properties from mixed fast and slow mode ultrasonic signals," J. Acoust. Soc. Am. **121**, EL8–EL15 (2007).
[13] K. Wear, "Decomposition of two-component ultrasound pulses in cancellous bone using modified least squares Prony method – Phantom experiment and simulation," Ultrasound Med. Biol. **36**, 276–287 (2010).
[14] A. Nowicki, J. Litniewski, W. Secomski, P. A. Lewin, and I. Trots, "Estimation of ultrasonic attenuation in a bone using coded excitation," Ultrasonics **41**, 615–621 (2003).
[15] M. Golay, "Complementary series," IRE Trans. Inf. Theory **IT-7**, 82–87 (1961).
[16] T. X. Misaridis and A. Jensen, "Use of modulated excitation signals in ultrasound, Part I: Basic concepts and expected benefits," IEEE Trans. Ultrason. Ferroelectr. Freq. Contr. **52**, 177–191 (2005).
[17] A. Nowicki, I. Trots, P. A. Lewin, W. Secomski, and R. Tymkiewicz, "Influence of the ultrasound transducer bandwidth on selection of the complementary Golay bit code length," Ultrasonics **47**, 64–73 (2007).
[18] B. Lashkari and A. Mandelis, "Linear frequency modulation photoacoustic radar: Optimal bandwidth for frequency-domain imaging of turbid media," J. Acoust. Soc. Am. **130**, 1313–1324 (2011).
[19] D. L. Johnson, T. J. Plona, C. Scala, F. Pasierb, and H. Kojima, "Tortuosity and acoustic slow waves," Phys. Rev. Lett. **49**, 1840–1844 (1982).
[20] A. Scheidegger, *The Physics of Flow Through Porous Media*, 3rd ed. (University of Toronto Press, Toronto, 1974).