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Technical note

Measurements of acoustic dispersion on calcaneus using spilt spectrum processing technique

Pei-Jarn Chen, Tainsong Chen *

Institute of Biomedical Engineering, National Cheng-Kung University, No. 1, Ta-Hsueh Road, Tainan 70101, Taiwan

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Abstract

The speed of sound (SOS) has become a useful tool in osteoporosis assessment, since it represents a combination of density and compressibility of bone tissue and should provide better information on bone quality and an estimate the fracture risk. In general, the speed of sound on dispersive material, such as bone tissue, depends strongly on frequency. Therefore, a measurement of velocity dispersion magnitude (VDM) might provide more important bone structure information than measurements of bone mineral density (BMD), SOS or broadband ultrasound attenuation (BUA). To obtain the velocity dispersion magnitude requires a sequence of pulses that have a frequency that is different from that used in conventional approaches. The measurement is complicated by the fact that pulse waveform will distort as the pulses propagate through the frequency-dependent medium. Alternatively, the phase velocity and velocity dispersion measurements also can be obtained on frequency-domain processing. However, the accuracy of those techniques is affected by the $2m\pi$ ambiguity in the phase unwrapping process in frequency domain. And the spectrum approach is highly dependent on the gating window selection in time domain signals. The time-domain split spectrum processing (SSP) technique is proposed here to measure the phase velocity and the VDM. The SSP technique is also used to measure the SOS and VDM of two commercial calcaneus phantoms. Simulation results are in good agreement with the preset parameters of a model-based signal obtained using the SSP technique. In addition, in vitro SSP measurements agree with the manufacturer's specifications for two commercial calcaneus phantoms. The negative dispersion is also found in in vivo measurements on human heel. Finally, an approach based on the time domain SSP technique has potential clinical applications for osteoporosis diagnosis.

Keywords: SSP; Velocity dispersion; Phase velocity; Ultrasound; Calcaneus

1. Introduction

Ultrasonic techniques have become popular bone assessment tools in recent years. The speed of sound (SOS) and broadband ultrasound attenuation (BUA) provide the information of bone quality such as bone density, bone micro-structure and bone elasticity [1,2]. Many studies have shown that there is a strong association between quantitative ultrasound (QUS) measurements taken at the heel and fracture risk, and that calcaneal QUS can be used to predict osteoporotic fractures [3,4], including hip fractures [5,6] and vertebral fractures [7–9] just as well as dual-energy X-ray absorptiometry (DXA). Therefore, there is a need to develop ultrasound measurement techniques that improve upon, and go beyond, the existing measurements of SOS and BUA for osteoporosis assessment.

Most of the techniques for estimation of SOS depend on the time-of-flight (TOF) measurements. Several methods have been proposed to obtain the TOF between two pulses including measurement of pulse envelop peak or zero-crossings. However, a broadband ultrasound pulse will be highly distorted as it propagates through a high attenuation medium such as calcaneus [10]. This distortion in pulse shape will result in erroneous TOF measurements when conventional approaches are used. In general, a high attenuation media like calcaneus bone is also highly dispersive [11]. In this situation, the dispersion will cause additional distortion of the pulse waveform and these will further degrade the accuracy

* Corresponding author. Tel.: +886 6 2757575x63425; fax: +886 6 2760677.
E-mail address: chents@mail.ncku.edu.tw (T. Chen).

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of time domain measurement of SOS. Ambiguities associated with these transit-time based methods for estimation of TOF in bone have been reported by several researchers [12,13]. At the same time, above approaches provide only the SOS at certain frequency. This problem has led some investigators to suggest the use of phase velocity, which is measured in the frequency domain, as an alternative to transit-time methods [12–14]. The dispersion of phase velocity was found to be virtually independent of the BMD in experiments by Wear [15]; however, if the velocity dispersion magnitude (VDM) is related to the histomorphometric properties of trabecular bone, as has been postulated by Droin [14], then the VDM should convey important information about bone structures, information not contained in either the BMD, SOS or BUA measurements. This motivates our efforts to develop a time-domain SSP technique to measure the velocity dispersion of bone tissue.

In general, a sequence of narrowband pulses with different frequencies is needed to obtain the dispersive properties of bone tissue [16]. Recently, the velocity dispersion has been evaluated by measuring the phase velocity over different frequencies in the frequency domain [12–14]. However, the accuracy of those techniques is affected by the $2m\pi$ ambiguity in the phase unwrapping process and the number of sampling points or the shift points of the central pulse when using the so-called circularly rotating techniques [11,17]. In addition, the spectrum approach is highly dependent on the gating window selection in time domain signals. Consequently, a time-domain theory proposed by Wear has been derived to compensate for the variations in transit-time based SOS estimates assuming a Gaussian pulse propagating through attenuated, weakly dispersive media [10,18]. This approach requires simplifying assumptions such as linear dispersion and a linear model of the medium. Nevertheless, the technique performs rather well based on experimental results.

The split spectrum processing (SSP) technique was introduced in the late 1970s to implement a frequency agility technique used in radar to improve signal-to-noise ratio (SNR) [19,20]. The SSP technique achieves frequency diversity by splitting the ultrasound signal into a number of signals each with different frequency components using a linear band-pass filter or non-linear frequency-diverse statistical filter [21]. In this approach, one broadband ultrasound pulse interrogation can represent a number of pulse interrogations with different frequencies. Therefore, we can consider the incident ultrasound signals as a combination of several narrowband components. The received ultrasound signal can then be separated into several narrowband components ultrasound signals using the same filters. In this paper, a prior Gaussian filter bank such as used in Wavelet analysis is developed in the time-domain where the filters have constant bandwidth. Then, the incident ultrasound signals and the received ultrasound signals that have travelled through the calcaneus are decomposed into a number of pulses of different frequencies. The conventional approaches for TOF measurement are then applied to each ultrasound pulse within a given frequency band. Finally, the acoustic dispersion of bone tissue can be obtained by combining the individual measurements.

2. Materials and methods

2.1. Theoretical development

An optimal band-pass filter is a summation of several narrow Gaussian filters, which have a plateau over the original spectrum ranging from the lowest and highest central frequencies, f_L, f_H as shown in Fig. 1. If the plateau amplitude is designed as a constant and the range between f_L and f_H covers both the original and the received signal spectrum, then the original signal spectrum, $Y(f)$, will not be modified by the band-pass filter transfer function $H(f)$ as indicated in Eq. (1):

$$Y(f)H(f) = Y(f) \tag{1}$$

where the $H(f)$ represents the transfer function of band-pass filter and is the summation of several narrowband Gaussian filters as shown in Fig. 1. The transfer function $H(f)$ can be described as:

$$H(f) = \sum_{i=1}^{n} H_i(f) \tag{2}$$

$$H_i(f) = A\, \mathrm{e}^{-(f-f_i)^2/2\sigma_f^2} \tag{3}$$

$$h_i(t) = A\, \mathrm{e}^{-t^2/2\sigma_t^2} \sin(2\pi f_i t) \tag{4}$$

where $h_i(t)$ represents the time-domain signals of the filters with zero-phase as shown in Fig. 2; A is the magnitude, σ_f denotes the bandwidth and $\sigma_t = 1/2\pi\sigma_f$ is the duration of the

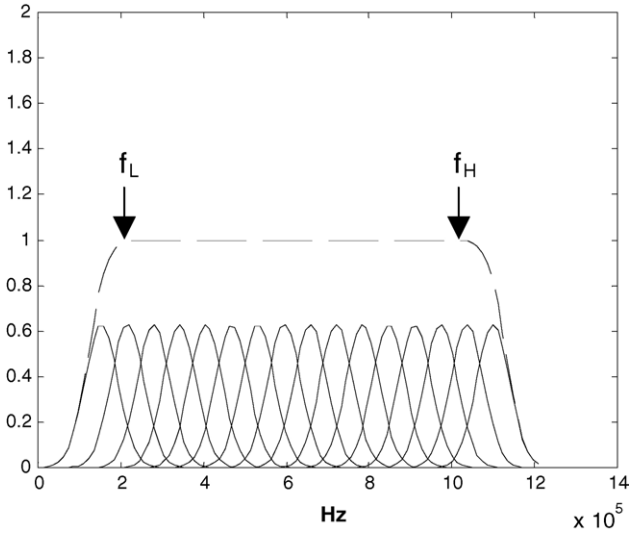

Fig. 1. The summation of the narrow-band Gaussian filters.

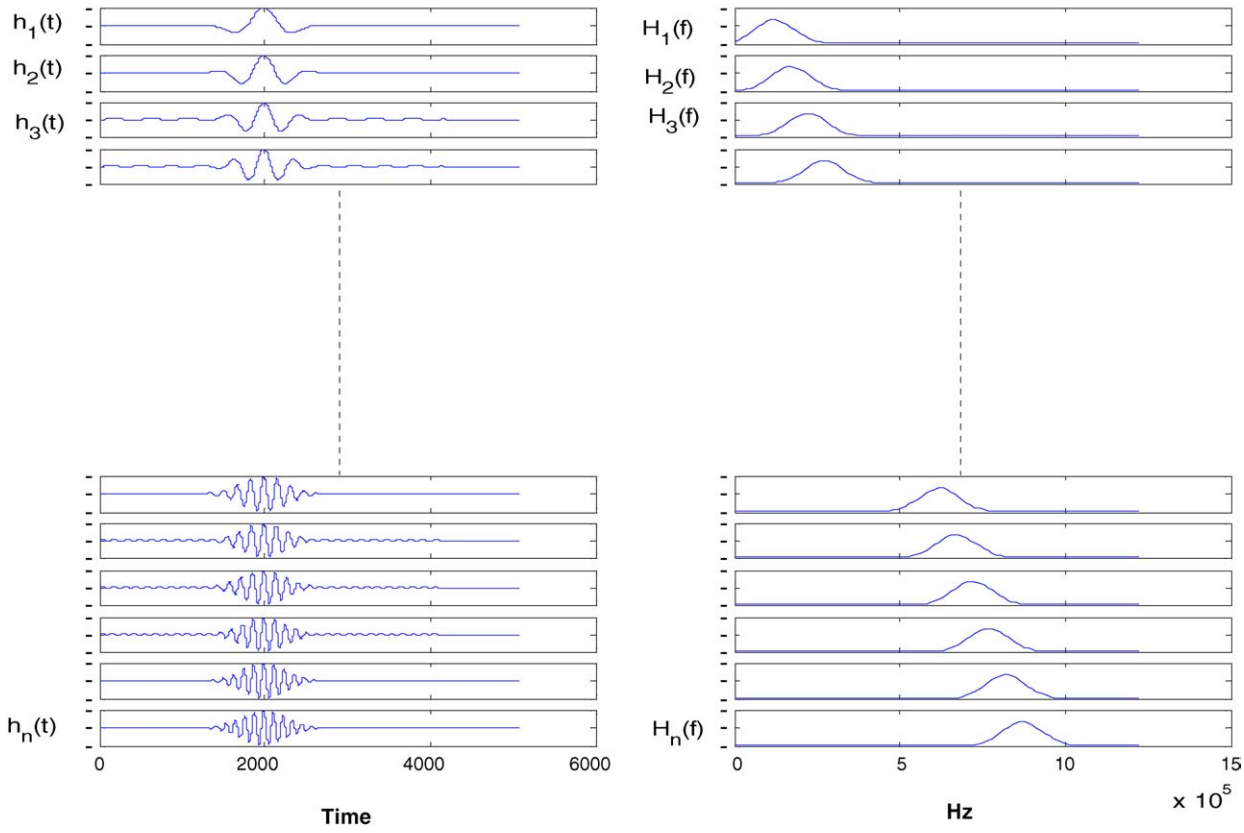

Fig. 2. The filter bank signals in time-domain (left) and the equivalent spectrum (right).

pulse. The time-domain Eq. (1) is shown in Fig. 3 and can be written as

$$y(t) \otimes \sum_{i=1}^{N} h_i(t) = \sum_{i=1}^{N} y_i(t) \tag{5}$$

This means that the original ultrasound signal, $y(t)$, can be decomposed into a number of signals, $y_i(t)$, with different nominal frequencies by convoluting $y(t)$ with each narrowband Gaussian filter, $h_i(t)$, in time-domain. The designed filter bank responses below f_L and above f_H are not constant and the original signal spectrum $Y(f)$ will be not conserved. The normalized root-mean-square (RMS) error calculated between the reconstructed signal, $y'(t)$, and the original signal, $y(t)$, will be large if the bandwidth of narrowband Gaussian filter is too large. Therefore, the bandwidth of each filter bank in our filter bank is given the same value and this value is chosen to be narrow enough so that the error between the downshift of each decomposed signal component and the total reconstructed RMS error will be negligible. This leads to a bandwidth criterion as follows [3,13]:

$$\sigma_f < \sqrt{\frac{2 f_0 \delta}{\beta \Delta d}} \tag{6}$$

where f_0 is the central frequency, β and Δd represent the attenuation coefficient and the travel length, respectively. The variable δ is a predetermined threshold parameter. As a result, the total number, N, of Gaussian filters give the total time of the signal, T, can be determined as [19,20]:

$$N = \sigma_f T + 1 \tag{7}$$

Many different filter bandwidths, σ_f, and total number filters, N, will satisfy both Eqs. (6) and (7), simultaneously. An optimal filter bandwidth (σ_f) will be determined only after choosing the minimum RMS between the original signal and the reconstructed signal. (The proposed filter bank is similar to those used in some wavelet analyses except that it is

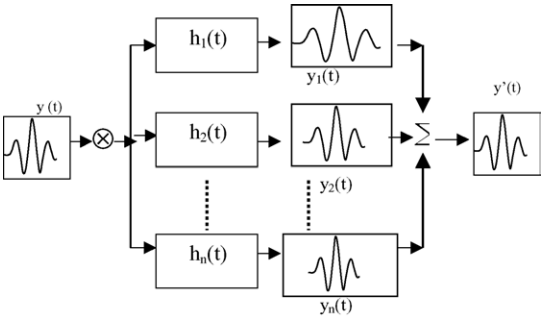

Fig. 3. The block diagram of signal decomposition.

constant bandwidth instead of constant Q.) Filter bank decomposition may be more suitable than other spectral methods (i.e. FFT, AR, ARMA, eigenvalue, etc.) because it is in the time-domain technique and provides a more intuitive and straightforward concept for clinicians involved in SOS measurements.

2.2. The experimental set-up

Two commercial bone phantoms were used to mimic normal bone tissue and osteoporotic bone tissue. Both phantoms are manufactured by CIRS Inc. (Norfolk, VA, USA). One was designed to mimic normal bone tissue (7.30 cm × 5.90 cm ($L \times W$), thickness = 3.63 cm, BUA = 69 dB/MHz, S0S = 1576 m/s). The other was designed to mimic osteoporotic bone tissue (7.30 cm × 5.90 cm ($L \times W$), thickness = 3.64 cm, BUA = 48 dB/MHz, S0S = 1501 m/s). Two unfocused broadband ultrasound transducers (1.5 in. diameters, Model #V389S-SU, Panametric Inc., Waltham, MA) with 0.5 MHz nominal central frequencies were used to measure the phase velocity using the SSP approach. An addition, three pairs of unfocused narrowband ultrasound transducers (1.0 in. diameter, TKS, Korea) with central frequencies of 0.25, 0.5 and 0.75 MHz were used to measure the group velocity at each frequency. These transducers were driven by a commercial ultrasound instrument (Panametric 5058 pulser/receiver, Waltham, MA). The specimen and transducers were tested in a water tank. To reduce the temperature effects on acoustic velocity, the temperature of the coupling medium was kept at 25 ± 0.1 °C during measurement. The received signals were acquired from a Tektronic 520D oscilloscope at a 100 MHz sampling rate with 8 bit resolution. The average of six measurements was used to reduce the signal noise level, and the signals were stored into a PC-type computer for further off-line processing.

3. Results

3.1. Filter design

The optimal bandwidth for each narrow-band Gaussian filter was selected as 40 kHz based on Eq. (6) and with a threshold parameter of $\delta = 5\%$. Fig. 2 shows the narrow-band Gaussian filter signals in the time (left) and frequency domain (right). The decomposition ranged from 0.1 to 1.1 MHz and the frequency plateau was constant between 0.2 and 0.8 MHz. The normal root-mean-square error between a simulated ultrasound pulse with Gaussian waveform and the signal reconstructed from the combination of narrow-band signals was less than 0.01% with our filter bank.

3.2. The simulation of model-based ultrasound signals

To validate the feasibility of acoustic dispersion measurements using the SSP approach, a Gaussian-shaped pulse having $f_0 = 0.5$ MHz, and $\sigma_f = 100$ kHz passing through calcaneus bone tissue (thickness =2 cm) was simulated. The parameters used in this simulation were typical for calcaneus: an attenuation coefficient of β was set to 20 dB/cm MHz; the SOS of water C_w was set to 1482 m/s; and the group velocity of ultrasound on calcaneus at the central frequency f_0 (0.5 MHz) C_{s0} was set to 1550 m/s [15,16]. The dispersion of acoustic speed, C_i, can be represented by following equation:

$$C_i = C_{s0} + b_s(f - f_0) \tag{8}$$

where the b_s, is assumed to be dispersion constants and ranged from −80 to +80 m/s MHz. A number of narrow-band signals with different nominal frequencies can be obtained from the incident or received ultrasound signals by convolution with the designed filters. Then the phase velocity, V_p, at each nominal frequency can be estimated directly using

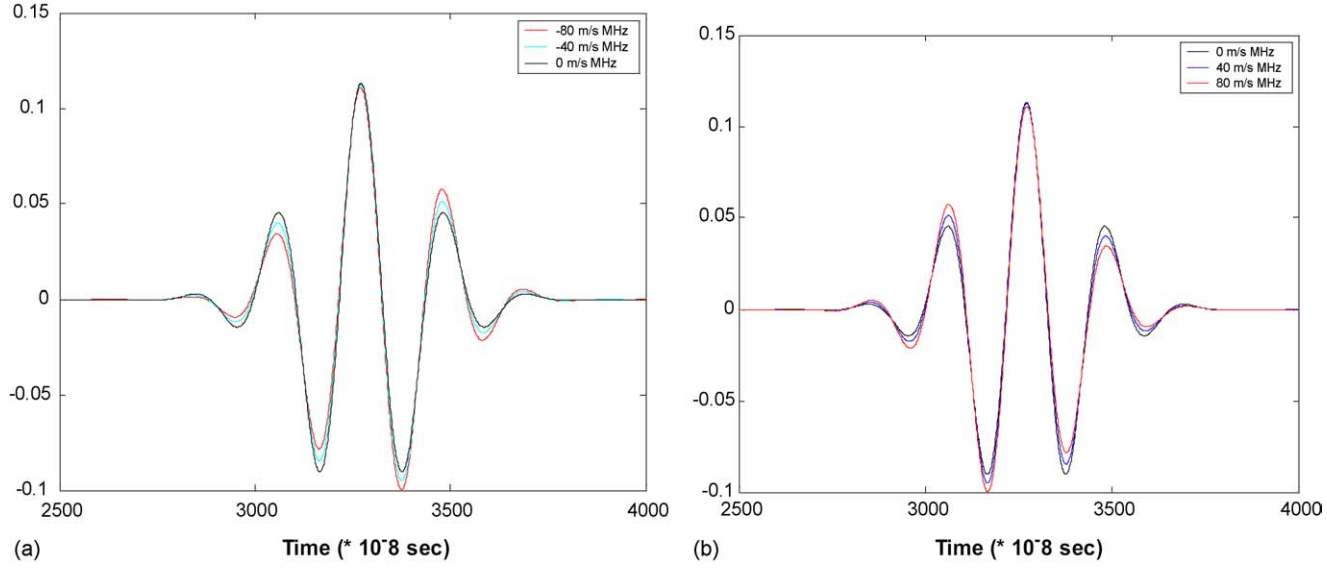

Fig. 4. The waveforms of ultrasound signal pass through calcaneus with different positive velocity dispersion parameters: (a) −80 to 0 MHz; (b) 0 to 80 MHz.

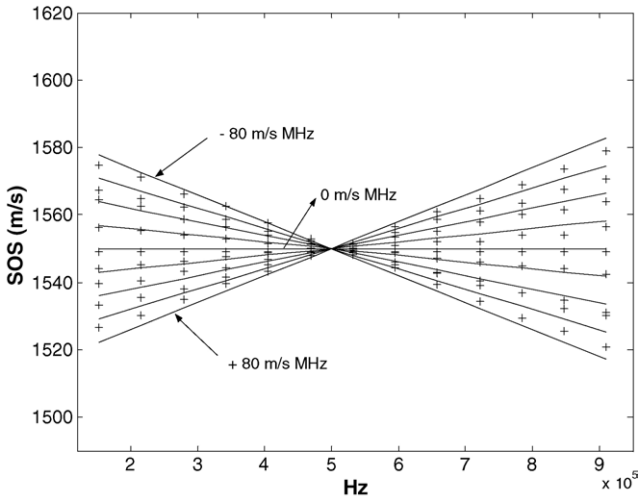

Fig. 5. The phase velocity obtained by SSP techniques with different velocity dispersions (−80 to 80 m/s MHz). (−) Presetting value with Eq. (10); (+) SOS obtained by SSP techniques.

conventional techniques either reflection mode (Eq. (9)) or substitution mode (Eq. (10)).

$$V_{\mathrm{p}}(i) = \frac{2\Delta d}{\Delta t_i} \tag{9}$$

$$V_{\mathrm{p}}(i) = \frac{\Delta d C_{\mathrm{w}}}{\Delta d - C_{\mathrm{w}}\Delta t_i} \tag{10}$$

The Δt_i represents the TOF differences between the incident and the received signals at ith frequency component signal. The numerical simulation programs were developed using MATLAB (MathWorks Inc., Natick, MA).

Signals with different dispersion parameters (b_{s}, ranging from −80 to 80 m/s MHz) are illustrated in Fig. 4. Negative velocity dispersion means that the high frequency components will propagate slower than the low frequency components, and the waveform will be distorted. These results reveal that the trailing edge of the pulse will be enlarged under negative dispersion. Conversely, the low frequency components will travel slower than the high frequency components when positive dispersion exists, and the leading edge will be larger than the trailing edge. These results are consistent with previous reports [10]. Consequently, the VDM is not easy to obtain using the time-domain techniques such as peak tracing and transit-time [12–14].

The results of the phase velocity obtained by the SSP techniques with both negative and positive velocity dispersion are illustrated in Fig. 5. These results show that the velocity dispersions obtained by the SSP techniques are consistent within the spectrum ranging from 0.2 to 0.8 MHz. The errors calculated from the preset VDM and the VDM measured by SSP technique are small less than 10%. These results demonstrate the feasibility of using the SSP technique when applied to model-based ultrasound signals.

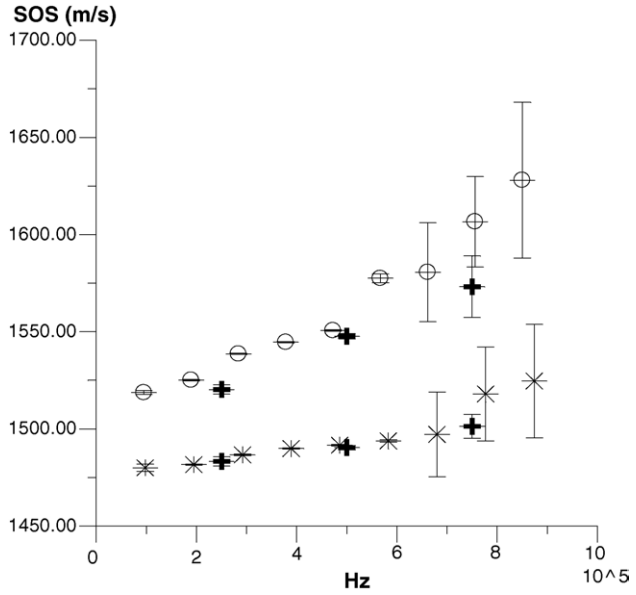

Fig. 6. The phase velocity measured by SSP technique with one broadband pulse and three different pulses with different nominal frequencies on osteoporotic bone phantom (*) and normal bone phantom (◯) (✚: 0.25, 0.5, 0.75 MHz).

3.3. In vitro and in vivo measurements

Fig. 6 illustrates the velocity dispersion measured by the SSP approach with one broadband signal and three different pulses with different nominal frequencies on osteoporotic and normal bone phantoms. The group velocities are all measured using peak trace techniques. The results show that the phase velocity estimated at 0.25, 0.5 and 0.75 MHz by the SSP techniques are consistent with the results obtained from three different transducers with 0.25, 0.5 and 0.75 MHz nominal frequencies as well as the manufacturer's specifications.

The volunteer, one of the authors, was tested to measure the acoustic dispersion of human calcaneus using the new technique. Six measurements were performed for both heels

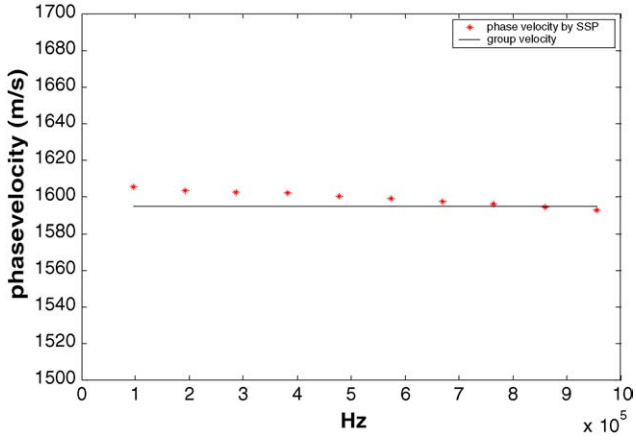

Fig. 7. The measurement of VDM on human calcaneus using proposed SSP technique.

at each test. Fig. 7 shows the result based on the proposed SSP technique. The negative dispersion is consistent with the results as reported by several researchers [12–15].

4. Conclusion and discussions

Sound speed is a fundamental property that has been measured in many soft tissues and conveys important information regarding tissue composition [22]. In addition to its importance to soft tissues, sound speed has been revealed to be highly correlated with calcaneal mass density [23]. Some researchers demonstrated that the SOS measurement from the time-domain could be highly dependent on the marker chosen within in the pulses [24]. The ultrasound pulse waveforms will be highly distorted when the pulse propagates through dispersive medium, such as bone tissue. Therefore, an error in the SOS estimation will occur when using current techniques. In general, most of the measurement techniques provide only the group velocity at a certain frequency. The bone tissue is a dispersive medium in which SOS is a function of frequency. Therefore, the VDM may provide more information on bone quality for osteoporosis assessment. The measurements of VDM need several interrogations with frequencies that are different than those used in conventional approaches.

In this paper, the SSP technique decomposes a broadband pulse into serial signals having different frequencies using a serial band-pass filters bank. In this way, the VDM can be easily obtained from one broadband ultrasound pulse with a single interrogation. The filter bank is similar to those used in wavelet analysis except that it is constant bandwidth instead of constant Q. Then if the incident ultrasound pulse does not have an ideal Gaussian shape, the received signal will be heavily distorted after the incident signal travels through the calcaeus phantom. As a result, it would be very difficult to determine the TOF using the conventional technique such as zero-crossing or the peak trace approach. The main advantage of the SSP technique is that we can easily determine the TOF between the peak position of the incident and received signals. This is because each subband signal will be more like a Gaussian waveform when the signals are convolved with the filters. The filter bank decomposition may be better than other spectral methods for measuring the VDM because it is a in the time-domain technique and will be more familiar to clinicians. However, the successful use of the SSP technique is highly dependent on an optimal design of the filter bank. The bandwidth of each band-pass filter should be narrow enough to meet the criteria described in Eqs. (6) and (7). Then, the summation of the all the band-pass filters will achieve a plateau in the frequency domain that covers the spectrum of the incident signal. In this case, the root-mean-square error between the incident signal and the reconstructed signal from the SSP technique will be negligible. The simulation results show that the VDM measurements obtained from the SSP technique are consistent with the preset model parameters. The results of the measurements on calcaneus phantoms are also in good agreement with manufacturer's specifications. In addition, the negative velocity dispersion revealed on human calcaneus measurements from the SSP technique is consistent with the reports estimated from in vitro measurements [12–15]. This suggests that the proposed SSP technique has potential for clinical application for osteoporosis diagnosis, since it is both simple and straightforward.

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