Improved standardization methods for clinical measurements of BUA and SOS

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This study is the first investigation of phase insensitive (PI) broadband ultrasound attenuation (BUA) on a clinically-relevant population with a clinical bone sonometer. Phase cancellation effects are significant for conventional phase sensitive (PS) BUA and can be mitigated using PI detection. Because of superior accuracy and lower coefficient of variation, PI processing is a promising alternative to PS processing for BUA measurements. In addition, a method for improving standardization of speed of sound (SOS) measurements is validated in the same clinical data set (73 women).

1 Introduction

1.1 BUA

Broadband ultrasound attenuation (BUA) is useful for prediction of fracture risk [1], but improvement in measurement methodology could increase its role in clinical practice [2]. The purpose of this study was to compare the conventional method, phase sensitive (PS) detection, with an alternative method, phase insensitive (PI) detection, for BUA measurements. PS reception takes the spatial integral (over the receiver aperture) of the *pressure field* incident upon the receiver. When some parts of the beam are out of phase with other parts, phase cancellation artifact results. PI reception takes the spatial integral of the *magnitude of the pressure field* incident upon the receiver. Therefore, PI reception is less prone to phase cancellation artifact [3].

1.2 SOS

Calcaneal speed of sound (SOS) is useful for prediction of fracture risk [1], but improvement in measurement methodology could increase its role in clinical practice [2]. SOS is usually measured with two transducers in a "pitch-catch" orientation. First, a reference measurement is performed by propagating a pulse from the transmitter to the receiver through a water path. Then the foot is placed between the two transducers, and a second measurement is performed. SOS is computed from

$$SOS = \frac{c_w}{1 + \frac{c_w \Delta t}{d}} \quad (1)$$

where c_w = acoustic velocity in water, Δt = difference in transit times of the two pulses, and d = bone thickness.

Different investigators choose different transit-time markers, as shown in Table 1. Variability in marker location, however, leads to variability in Δt, which leads to variability in SOS. See Figure 1.

The goal of this study was to test a compensation formula that corrects for the dependence of SOS measurements on experimental parameters [4]:

$$SOS_n - c_g \approx -\frac{\tau_n c_g^2 \sigma_f^2 \beta}{f_0^2} \frac{1}{1 - (\sigma_f^2 \beta d / f_0)} \quad (2)$$

where τ = (time between marker and pulse center) / waveform period, c_g =group velocity (SOS based on envelope maximum marker), f_0 = center frequency, σ_f = spectral standard deviation, BW = fractional bandwidth = σ_f / f_0, d = bone thickness, and BUA = broadband ultrasound attenuation.

Marker	Location	Author(s)
Leading edge		Njeh *et al.,* 1996, 1997
		Njeh & Langton, 1997
Thresholding	3 X noise std. dev.	Alves *et al.*, 1996a
	20% of 1st half cycle	Hakulinen *et al.*, 2005
Zero crossings	First	Nicholson *et al.*, 1998
		Trebacz & Natali, 1998
		Lee *et al.*, 2003
		Deligianni, 2007
	"Specific"	Rossman *et al.*, 1989
	Of first negative slope	Zagzebski *et al.*, 1991
	First after 10% threshold	Haiat *et al.,* 2006
	First after 15% threshold	Haiat *et al.*, 2005
Maximum absolute value		Alves *et al.*, 1996b
Maximum envelope		Wear, 2000.

Table 1. Transit-time marker locations used by various authors.

2 Methods

2.1 BUA and SOS

Through-transmission data were acquired in 73 women (mean age: 47 years, standard deviation: 13 years) using an Achilles Insight ® (GE Lunar, Madison, WI) bone sonometer. Radio frequency (RF) data from the receiver array were digitized and processed off-line using both PI and PS algorithms. See Figures 2 and 3.

Calcaneal bone mineral density (BMD) was measured on all 73 women using a GE PIXI dual-energy x-ray absorptiometer.

2.2 BUA

A phantom experiment was also conducted. A bone-mimicking phantom (GE Lunar, Madison, WI) was placed in a water tank between the transmitter and the receiver of the Achilles Insight. In order to simulate the uneven thickness of the calcaneus, an acrylic wedge was also placed in the acoustic path. The speed of sound in acrylic (2727 m/s) is faster than that in water (1480 m/s). See Figures 4 and 5.

2.3 SOS

A bone-mimicking phantom (Model 063, CIRS inc., Norfolk, VA) was interrogated using 1) the Achilles and 2) a Panametrics (Waltham, MA) 5800 pulser/receiver with a pair of coaxially-aligned Panametrics transducers.

"Uncompensated" SOS values were computed from Equation 1 for each of the following transit time markers: L3, L2, L1, T1, and T2 (τ = -5/4, -3/4, -1/4, 1/4, and 3/4). "Compensated" SOS values were computed by taking the uncompensated SOS values and subtracting the differences from group velocity predicted by Equation 2.

3 Results

3.1 BUA

Figure 6 shows ΔnBUA (excess normalized BUA compared with the θ = 0º measurement for nBUA) for PS and PI processing as functions of wedge deflection angle (θ).

The PI measurements show far less variation with θ and therefore were a much more reliable indicator of the volume of acrylic intercepting the ultrasound beam (which was constant for all 5 wedges).

BMD T-scores ranged from -3.2 to 2.5 (mean: -0.4, standard deviation: 1.1). BUA measurements were 81 ± 21 dB/MHz (PS) and 67 ± 10 dB/MHz (PI). Therefore, phase cancellation artifact accounted for 14 dB/MHz on the average. Coefficients of variation were 26% (PS) and 14% (PI). See Figure 7.

3.2 SOS

Figure 8 shows that compensated measurements (o's) were more consistent than uncompensated measurements (x's) in vivo. The trend for uncompensated measurements is like that reported by Laugier *et al*. [4].

Figure 9 shows that the Panametrics system had a wider fractional bandwidth (BW = σ_f/f_0) than the Achilles (25% vs. 17%). The wider-bandwidth system (Panametrics) showed greater variability of uncompensated SOS in a phantom (Figure 10) and the compensation formula suppressed variability of SOS (Figure 11). At L3 (a typical marker), a system difference (Achilles vs. Panametrics) of 41 m/s before compensation was reduced to only 5 m/s after compensation.

4 Conclusion

This study is the first investigation of PI BUA on a clinically-relevant population with a clinical bone sonometer. Phase cancellation effects are significant for conventional (PS) BUA and can be mitigated using PI detection. Because of superior accuracy and lower coefficient of variation, PI processing is a promising alternative to PS processing for BUA measurements.

This SOS compensation method can be used to improve consistency in bone sonometry by reducing the dependence of SOS estimates on transit-time marker location, bandwidth, and other experimental parameters.

Acknowledgments

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References

[1] D. Hans *et al.,* "Ultrasonographic heel measurements to predict hip fracture in elderly women: the EPIDOS prospective study," *Lancet*, 348, 511-514, 1996.

[2] C. C. Glüer, "Quantitative ultrasound—It is time to focus research efforts," *Bone*, 40, 9-13, 2006.

[3] L. J. Busse and J. G. Miller, "Detection of spatially nonuniform ultrasonic radiation with phase sensitive (piexoelectric) and phase insensitive (acoustoelectric) receivers," *J. Acoust. Soc. Am.* 70, 1377-1386, 1981.

[4] K. A. Wear, "The dependence of time-domain speed-of-sound measurements on cnter frequency, bandwidth, and transit-time mariker in human calcaneus *in vitro*," *J. Acoust. Soc. Am.*, 122, 636-644, 2007.

[5] P. Laugier *et al.*, "Ultrasound images of the os calcis: a new method of assessment of bone status," *Proc IEEE Ultrason. Symp. 2*, 989-992, 1993.

[6] K. A. Wear, "The effect of phase cancellation on estimates of calcaneal broadband ultrasound attenuation *in vivo*," *IEEE Trans. Ultrason., Ferro., and Freq. Contr.*, 54, 1352-1359, 2007.

[7] K. A. Wear, "The effect of phase cancellation on measurements of attenuation in human cancellous bone *in vitro,*" *IEEE Trans. Ultrason., Ferro., and Freq. Contr.,* 55, 384-390, 2008.

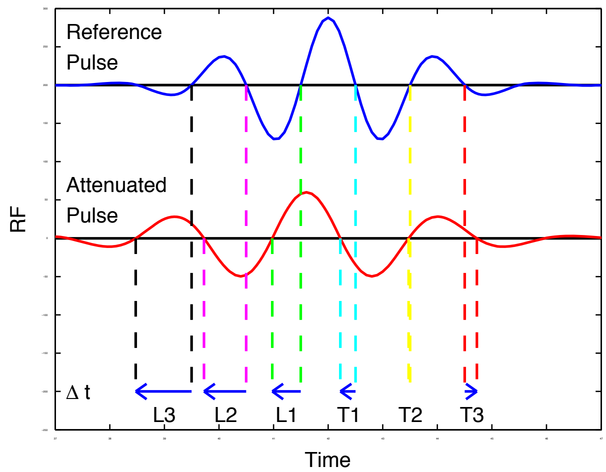

Figure 1. Since the attenuated pulse is low-pass filtered (stretched in time) due to frequency-dependent attenuation of the bone, Δt varies with marker location. Transit-time markers are labeled L3, L2, L1, T1, T2, and T3 where L and T denote the leading and trailing halves of the pulse and markers are numbered outward from pulse center.

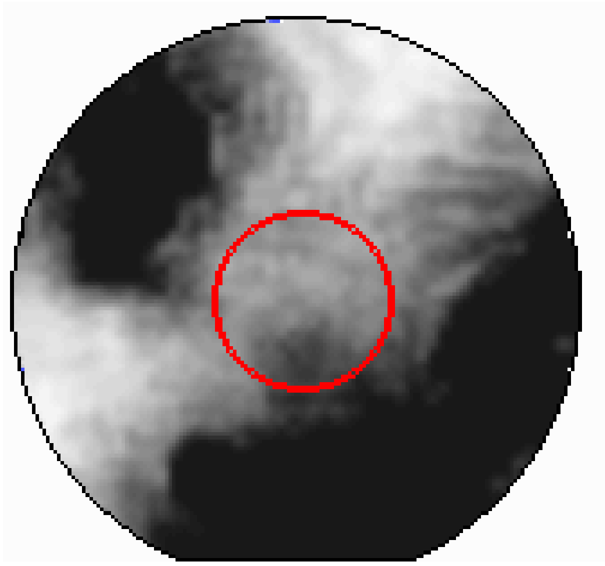

Figure 2. BUA image of calcaneus.

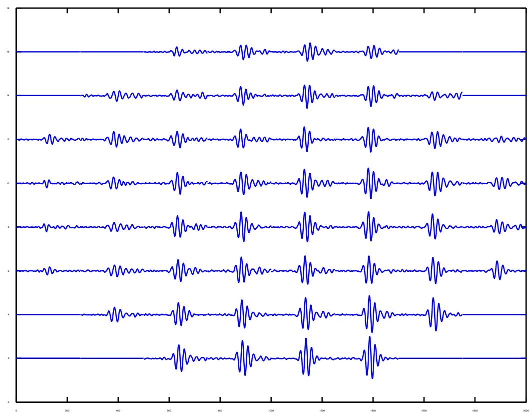

Figure 3. 52 RF signals corresponding to circled region of interest in Figure 1.

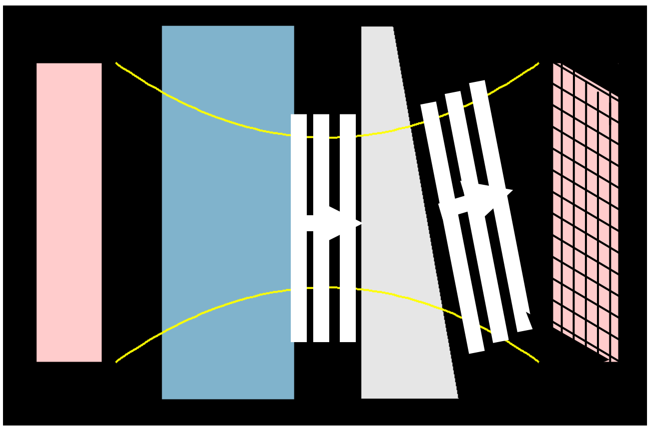

Figure 4. From left to right: 1) transmitter, 2) bone-mimicking phantom, 3) acrylic wedge, 4) receiver. The wedge deflects wavefronts, resulting in phase cancellation.

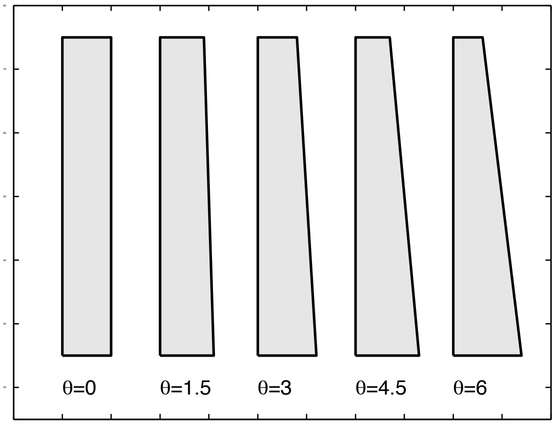

Figure 5. Acrylic wedges

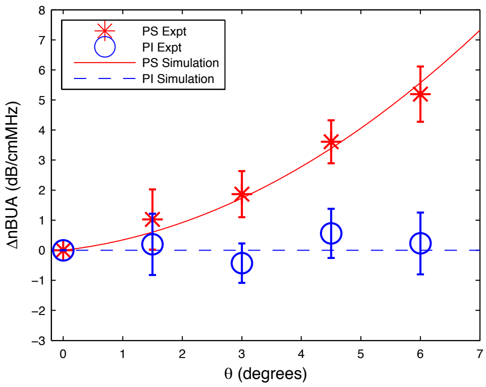

Figure 6. Acrylic wedge results.

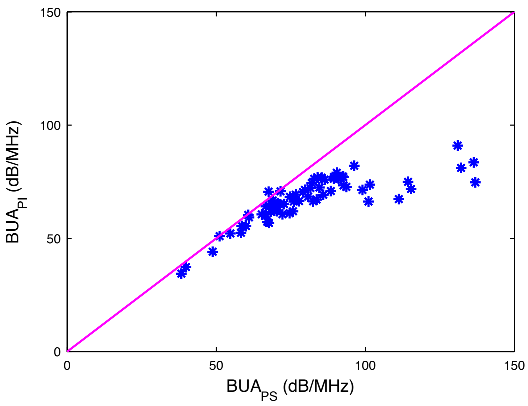

Figure 7. Clinical BUA results

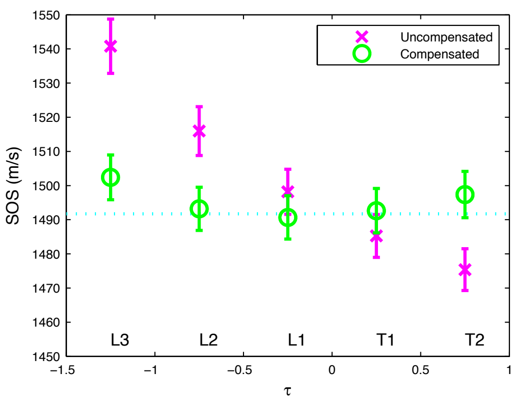

Figure 8. Clinical SOS results

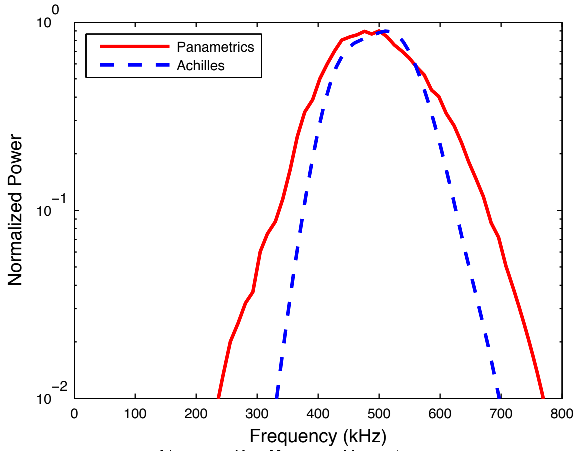

Figure 9. Power Spectra

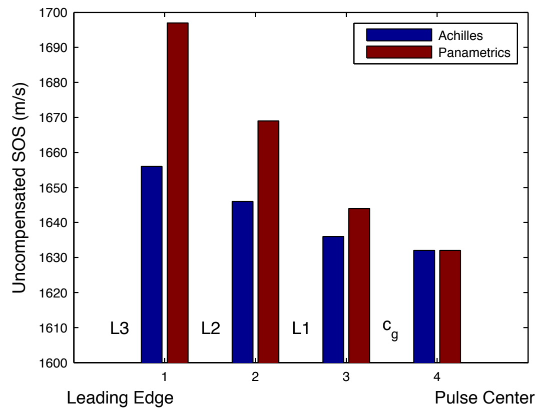

Figure 10. Uncompensated phantom SOS measurements

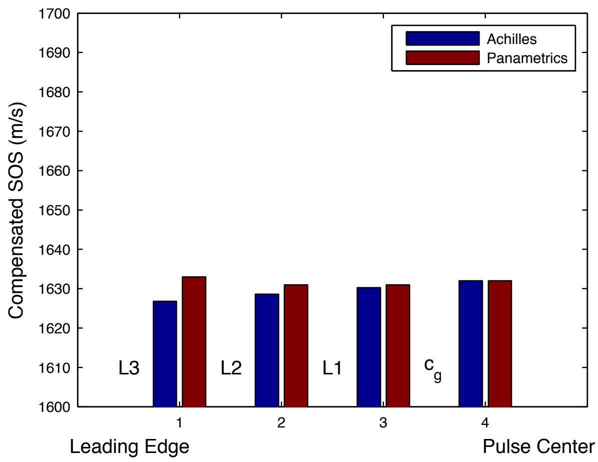

Figure 11. Compensated phantom SOS measurements.