Comparison of Techniques for *In Vivo* Attenuation Measurements

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Abstract—**The attenuation of an ultrasound pulse within tissue can be estimated from either the amplitude decay or the frequency downshift of returning echoes. This paper compares the results of both analyses applied to ultrasound *B*-scan echoes from the livers of 49 individuals. The amplitude decay of the backscattered signal Fourier components with depth was used to calculate attenuation coefficients. In addition, the frequency downshift of the same backscattered signals was estimated using both zero-crossing and spectral centroid methods. The analysis employed multiple regions of interest, each approximately 5 × 4 cm in dimension, from one or more liver *B* scans of each individual. The results show that the frequency-domain estimators yield consistently higher attenuation coefficients, with higher variability compared to the amplitude decay method. Explanations for the apparent bias and variability of the frequency-shift estimators include the assumptions regarding tissue and signal which may not be met in practice, and the effects of low-frequency electronic noise on spectral estimates.**

INTRODUCTION

ONE APPROACH to ultrasonic characterization of tissue utilizes quantitative measurements of attenuation and backscatter coefficients of normal or diseased tissues. Attenuation measurements have been studied by a number of groups [1]–[15] using various techniques and different assumptions regarding the frequency dependence of attenuation and pulse-spectral shape. One group of techniques begins with an assumption that attenuation increases linearly with frequency

$$\alpha = \beta f \tag{1}$$

where the single-parameter β is used as a measure of attenuation. This model is limited in its ability to represent tissue attenuation [4], [16], and therefore leads to systematic, frequency-dependent errors [12]. Nonetheless, (1) has served as the starting point for a number of estimation techniques. One of the first approaches utilized a pairwise comparison of backscatter spectra from different depths within tissue. Using (1), the slope of a straight line fit to the log ratio of two spectra (or difference between log spectra) yields β [9], [10]. The pairwise comparison of spectra has been shown to be an inefficient use of data [5], and the technique apparently has not been utilized in recent work [13], [14]. However, a parametric approach uses the β assumption of (1), plus the assumption of Gaussian spectral shape of the propagation pulse and echo [10], [17]. The higher rate of attenuation of higher frequencies causes a downshift in center frequency with depth. The amplitude attenuation coefficient can be calculated by

$$\beta = \frac{\Delta \bar{f}(d)}{\Delta d 2\sigma^2} \tag{2}$$

where $f(d)$ is the center frequency of the backscattered echo at any depth d in tissue, and σ is the Gaussian bandwidth (one standard deviation) of the amplitude pulse spectra. The factor of 2 in the denominator accounts for the round trip travel of the echo. The slope $\Delta \bar{f}/\Delta d$ is usually estimated from a straight line fit to a plot of $\bar{f}(d)$ where the bar represents averaging over a number of scan lines from a region of interest in tissue.

A number of approaches are used to estimate $\bar{f}(d)$ from the data. In the time domain, it is possible to use the zero crossing estimate [8], [18], [20]. The number of sign changes (zero crossings) per unit interval gives an estimate of center frequency, under the assumption of narrow-band signal. Alternatively, the transform of the signal can be used to calculate the first moment

$$f(d) = \frac{\int f S_d(f)\, df}{\int S_d(f)\, df} \tag{3}$$

where $S_d(f)$ is the magnitude of the amplitude spectrum from depth d, limited to finite time (finite sample volume), using an appropriate window.

An alternative approach uses the decay of the backscatter Fourier components with depth to estimate attenuation as a function of frequency [1], [2], [5], [7], or decay of the RF envelope to estimate attenuation at the center frequency [1], [3]. No *a priori* assumptions regarding the frequency dependence of attenuation, the transmitted spectral shape, or the scattering frequency dependence are required, and the analysis is not restricted to narrow-band signals. The amplitude decay technique has been shown to yield accurate and reproducible results on phantoms [1], provides consistent results for attenua-

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tion of normal livers [2], [7] closely matching the attenuation coefficients of excised mammalian livers [4], [16], [19], and has been useful in tissue characterization and attenuation imaging applied to normal and diseased livers [3], [7]. In this paper, the attenuation estimates derived from amplitude decay are compared with the parametric estimations of β, using data from *B*-scan liver examinations of 49 individuals.

Methods

The ultrasonic imaging system used in our study is the Octoson (Ausonics, Inc., Australia). In this system, the acoustic energy is transmitted and received by up to eight large aperture, wide-band transducers having a center frequency which we found to be near 2.5 MHz. The transducers are arranged in arc at the base of a deep-water path chamber and are focused at a range of approximately 35 cm. Ultrasonic echo signals are amplified by a wide-band receiver which has a nonlinear compression characteristic and time-varying gain. The amplified RF waveforms are transmitted to both the Octoson imaging circuits and to an 8 bit, 10 MHz, A/D converter for storage in digital form and later analysis. The time-varying gain values are also stored for processing which removes gain and compression effects. This processing is accomplished by using measured input–output characteristics of the Octoson [1].

Ultrasonic *B*-scan examinations were performed on 49 individuals including healthy volunteers and patients with documented liver disease. Multiple sagittal and transverse scans were digitized in most cases. To begin signal processing, speckle regions free from specular reflection and vessels, of size 4 cm wide and no longer than 5 cm in depth were selected from *B*-scan images for processing to determine absolute attenuation. A representative liver scan with region of interest is shown in Fig. 1. Then, 100 individual waveforms from these regions were isolated and the effects of receiver compression and time-varying gain were removed. Next, the waveforms were divided into segments of 128 data points. At the 10 MHz sampling rate, this corresponds to a distance of approximately 9.6 mm. Depth increments of 28 points were employed. An FFT routine was used with Blackman windows to calculate the spectral magnitudes as a function of frequency and depth. Alternatively, zero-crossing analyses were performed on unwindowed 128 sample segments to estimate local center frequency. Averaging across waveforms was performed in each analysis to produce a curve of 1) $\bar{f}(d)$ obtained from zero crossings, 2) $\hat{f}(d)$ obtained from spectral centroid estimates, and 3) $S_d(f)$ as a function of depth.

Diffraction corrections to these data were small due to the unusually large path length of the Octoson. Measurements were taken near the focal region, 35 cm $\pm$ 2.5 cm from the transducer. From our phantom and calibration studies [1], [2], amplitude corrections in this range are found to be less than 10 percent and diffractive frequency shifts less than 20 KHz. Thus, slight amplitude corrections but no frequency corrections were applied to the data. Analyses of attenuation utilized least-square error curve fits to curves 1) and 2), with β determined by (2). Fits of exponential decay to curve 3) at each discrete frequency between 2 and 3 MHz provided attenuation coefficients as a function of frequency. These were curve fit to a power law, and the value at center frequency 2.5 MHz divided by 2.5 MHz was used as single parameter to be consistent with β estimates. Analyses of any region of interest was disregarded if the curve-fit procedures yielded a correlation coefficient of less than 0.8, indicating poor match of data to the models.

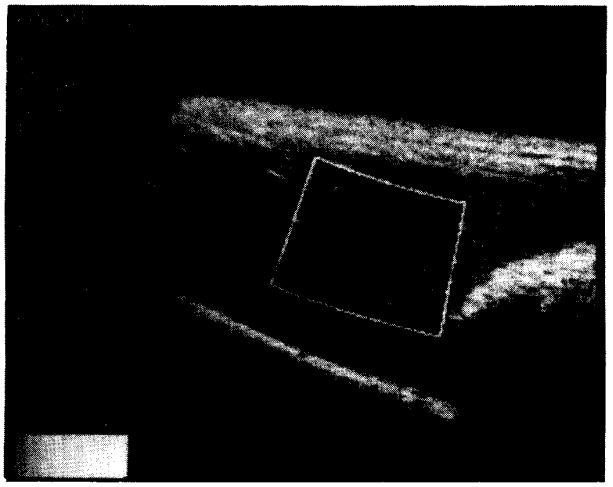

Fig. 1. Representative abdominal *B*-scan image (sagittal plane) with region of interest in liver outlined.

Results

Over the entire group of liver scans, attenuation values using amplitude decay ranged from 0.25 to 0.90 dB/MHz · cm, with a typical standard deviation between regions of interest from the same liver of approximately 0.1 dB/cm · MHz. In comparison, spectral-shift measurements on the same data resulted in a range of 0.18 to 1.55 dB/cm · MHz with an intraliver standard deviation of 0.2 dB/cm · MHz. The zero-crossing results were similar to the centroid shift, ranging from 0.25 to 1.55 dB/cm · MHz with an intraliver SD of 0.2 dB/cm · MHz.

Pairwise comparisons of results between the three techniques are given in Figs. 2, 3, and 4. Typical error bars are shown in the upper left quadrant of Figs. 2, 3, and 4 for the different estimators.

The data in Figs. 2 and 3 show that the parametric methods (centroid shift and zero crossing) yield consistently higher attenuation coefficients than does the amplitude decay method. The results in Fig. 4 show that of the two parametric methods, zero-crossing estimates tend to be higher than centroid shift estimates of attenuation.

Discussion

The parametric measurements of attenuation, using spectral centroid and zero-crossing estimates of center frequency, generally yielded high attenuation coefficients

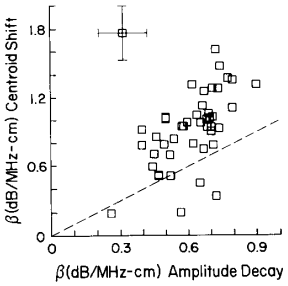

Fig. 2. Comparison of attenuation values estimated from amplitude decay and centroid shift. The dashed line is the line of identity and error bars in upper left represent typical intraliver variations.

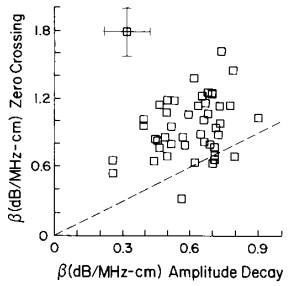

Fig. 3. Comparison of attenuation values estimated from amplitude decay and zero crossings. The dashed line is the line of identity and error bars in upper left represent typical intraliver variations.

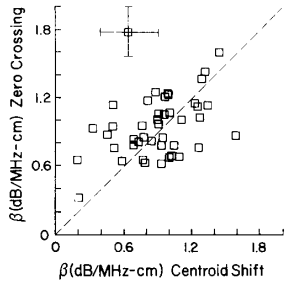

Fig. 4. Comparison of attenuation values, obtained from parametric methods. The dashed line is the line of identity and error bars in upper left represent typical intraliver variations.

with higher intra- and interliver variability than the amplitude decay techniques. The variability of parametric techniques has been noted in clinical measurements by others [14], [15]. Some reasons for the higher attenuation values and variability have theoretical bases while others have experimental bases which may be unique to the Octoson system that we used for *B*-scan examination.

Theoretically, if attenuation is well described by a power law model

$$\alpha(f) = \alpha_0 f^n \qquad n \neq 1 \tag{4}$$

then the linearity (β) assumption of the parametric methods is invalid. Specifically, a Gaussian spectrum is downshifted, but the downshift is related to α_0 and n and the bandwidth of the spectrum decreases with depth instead of remaining constant [20]. Normal mammalian liver tissue attenuation is well described by power law functions in the low MHz band, with n of 1.1–1.3 (2, 4, 7, 16, 19). Abnormal livers and other tissues have a wider range of n (2, 4, 7, 11); thus, linearity assumptions may to be of limited applicability to tissues. For cases where $n > 1$, application of the parametric techniques will lead to positive bias in attenuation estimates [12], [20].

Other factors which influence the results are the spectral bandpass shape and noise characteristics of the system. These factors are illustrated in Fig. 5. Fig. 5(a) shows received, ensemble average backscatter spectral magnitudes from different depths within a normal liver. Despite the averaging over large regions, curve-fit procedures show these spectral shapes do not converge to a Gaussian. The decay of amplitude at 2.5 MHz is given in Fig. 5(b), and this decay requires no assumptions about spectral shape to analyze. Fig. 5(c) shows the spectral centroid and zero-crossing estimates of frequency, along with straight line fits. Fig. 5(d) is the spectral bandwidth estimate, which increases with depth.

Theoretically, the Gaussian spectral bandwidth should remain constant under the linearity assumption, or decrease under the power law $n > 1$ assumption. In our system, the increased bandwidth most likely resulted from low frequency (1–2 MHz) electronic noise. Independent measurements of the system amplifiers showed higher electronic noise at high time-gain-control values, which in practice are correlated with deeper, attenuated signals. Thus, the lower signal-to-noise ratio with depth would result in broader spectra and the trend shown in Fig. 5(d). Since the electronic noise was most prominent at low frequencies (1–2 MHz in this system), this would bias the center-frequency estimates downward with depth, and therefore bias parametric attenuation estimates upward. In contrast, since the amplitude decay method utilized Fourier components between 2–3 MHz, results from this method are relatively unaffected by any increase in the low-frequency amplifier noise. Therefore, some results of our comparisons depend on the characteristics of the particular instrument used, and cannot be generalized without specific noise and time-gain-control measurements of other systems.

Conclusion

In vivo liver attenuation estimates were obtained from digitized RF *B*-scan echoes of 49 individuals. Amplitude decay of Fourier coefficients with depth was compared against parametric techniques based on zero crossing and spectral shift estimators. Parametric estimations were found to yield higher and more variable values of liver attenuation coefficient. The results may be explained partly in terms of the backscatter spectral shape and frequency dependence of attenuation assumptions which are not met in practice. Also, low-frequency noise at high-amplifier gain would affect center frequency estimates but not amplitude decay estimates.

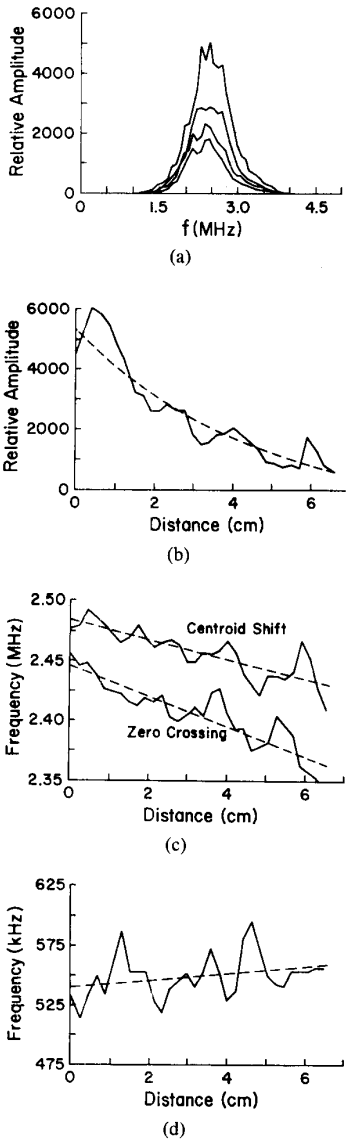

Fig. 5. Representative data. (a) Average spectral magnitudes from four depths within liver. The data show a generally decreasing amplitude with increasing depth. (b) Decay of average pressure amplitude at 2.5 MHz, versus depth within liver region of interest. The dashed line is least square error fit of an exponential to the observations. (c) Frequency shifts versus depth using centroid estimator (top curves) and zero crossing. The dashed lines are least-square error fits to a linear frequency shift with depth. (d) Bandwidth as measured by standard deviation of averaged spectra versus depth. The dashed line is a least-square error fit to a linear function. Almost all cases showed a slight but consistent increase in bandwidth with increasing depth, even though theory predicts a constant or decreasing bandwidth.

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