Phase insensitive ultrasonic attenuation coefficient determination of fresh bovine liver over an extended frequency range[a)]

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The ultrasonic attenuation coefficient of fresh bovine liver was determined by the phase insensitive techniques of the radiation force balance at 1.37, 4.15, 6.90, and 9.65 MHz, and the scanning laser acoustic microscope (SLAM) at 100 MHz. A least squares fit of the ultrasonic attenuation coefficient (in Np/cm) vs ultrasonic frequency (in MHz) to the 1–10 MHz data yields $A = 0.043 f^{1.266}$ ($r**2 = 0.895$) and to the 1–100 MHz data yields $A = 0.043 f^{1.270}$ ($r**2 = 0.954$). The following observations can be made: the greatest frequency range (2 decades) over which ultrasonic frequency has been determined for the same specimens. (2) This is the first time it has been shown that the frequency dependence of attenuation remains essentially unchanged over this large frequency range. (3) The frequency dependence is slightly greater than has been observed from the various literature compilations. (4) The attenuation coefficient magnitudes are the lowest values yet determined for fresh liver. (5) The attenuation magnitudes are from 1.3 to 2.2 times greater than the ultrasonic absorption values in the same tissue.

Key words: ultrasound, attenuation coefficient, frequency dependence

INTRODUCTION

In order to study the mechanisms involved in the propagation of ultrasound through tissue, it is necessary to obtain data which have been measured under well documented and reproducible conditions over a wide range of frequencies. Additionally, since fixatives appear to alter the ultrasonic propagation properties, these measurements should be made in fresh tissues to more closely represent *in vivo* conditions. Examination of recent compilations[1,2] have shown that such a data base is not available. This study was undertaken to obtain such a data base. Bovine liver tissue was chosen because of its availability.

METHODS

All measurements reported herein were performed within a few hours postmortem on fresh bovine liver specimens at room temperature (22°–25°C). The liver specimens were obtained from a slaughter house and were transported to the laboratory in normal saline solution at room temperature (22°–25°C). They were then immediately divided into sections for simultaneous measurement by both attenuation measurement methods. The tissues were also bathed in the saline solution during the actual measurement process.

The radiation force balance technique is a phase insensitive, insertion loss technique for measuring the ultrasonic attenuation in tissues. A highly sensitive chemical balance is used to measure the radiation pressure of an ultrasonic beam impinging on an absorbing target suspended from the balance beam. Since the target is designed to intercept the entire ultrasonic beam, the radiation pressure is directly related to the power of the impinging ultrasonic signal. By determining the ultrasonic power, both with and without a tissue sample inserted into the ultrasonic beam, the amplitude attenuation coefficient of the sample can be determined from the expression:

$$A = -[\ln(P/P_0)]/2x, \tag{1}$$

where P and P_0 are the ultrasonic power with and without the tissue in the ultrasonic beam, respectively, and x is the thickness of the tissue sample. This procedure would include reflective losses at the tissue–saline interfaces in the value of attenuation coefficient. However, every effort was made to impedance match the saline to the tissue in order that reflective losses would be negligible and this was confirmed by measuring various thicknesses of the same specimen.

The radiation force balance apparatus employed in this study was developed by Nider,[3] and is similar to systems described by Rooney[4] and Carson *et al.*[5] It employs an absorbing target constructed of sound absorbing rubber (SOAB), and a Cahn electrobalance (model RG). The ultrasonic power levels employed ranged from 93 μW to 0.39 mW, and the specimen thicknesses ranged from 0.95 to 1.43 cm. The attenuation coefficients had maximum uncertainties of ±0.024 Np/cm at 1.37 MHz, ±0.044 Np/cm at 4.15 MHz, ±0.102 Np/cm at 6.90 MHz, and ±0.106 Np/cm at 9.65 MHz, and the specimen thickness had a maximum uncertainty of ±0.075 cm.

The scanning laser acoustic microscope (Sonomicroscope 100, Sonoscan, Inc., Bensenville, Il. 60106) also yields a phase insensitive measurement of attenuation. The reason for this is derived from the imaging technique employed in the SLAM, viz., dynamic ripple diffraction. This technique has been described in detail elsewhere.[6] A schematic diagram of

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TABLE I. Ultrasonic attenuation in fresh bovine liver.

Frequency (MHz)	Attenuation coefficient ± s.d. (Np/cm)	Number of liver specimens	Total number of measurements
1.37	0.069 ± 0.027	6	43
4.15	0.268 ± 0.077	5	32
6.90	0.477 ± 0.063	4	21
9.65	0.825 ± 0.205	2	12
100	15.4 ± 1.9	8	47

the SLAM image plane is shown in Fig. 1. The acoustic wavefronts are incident at an angle θ_0 (typically around 10° for biological specimens and their media) with respect to a normal to this liquid–solid interface.[7] The speed of propagation in the liquid is c_0, and the distance between wavefronts is λ_0, the acoustic wavelength. This creates a dynamic wave along the interface with a wavelength equal to $\lambda_0/\sin\theta_0$ (typically around 85 μm; c_0 = 1500 m/s), and traveling to the right at a velocity $c_0/\sin\theta_0$ (8500 m/s), as shown. When this liquid–solid interface is scanned with a focused laser beam from above (relative to Fig. 1), the dynamic wave causes the reflected laser beam to be angularly modulated at a frequency of c_0/λ_0, which is the same frequency (100 MHz) as that of the acoustic energy. The amplitude of the angular deflection of the reflected laser beam is proportional to the acoustic wave amplitude incident upon the image plane. Since the diameter of the laser spot which scans the image plane is approximately one fourth that of the surface wave's wavelength (20 μm and approximately 85 μm, respectively), the reflected laser beam is not subject to phase cancellation effects at the image plane. In practice, this "liquid" can be either the known, reference liquid or the unknown tissue specimen, but the above analysis is still approximately true since their propagation speeds are very similar.

Attenuation coefficient measurements are made with the SLAM in the following manner. An interference image of the particular specimen is obtained, and a portion of the TV screen is masked off, so that about one square inch of the screen in visible. This corresponds to an area on the specimen of approximately 0.13 mm^2. The masking is done so that the measurement in a particular region of interest is not affected by the visualization of the surrounding regions. The electrical, 100 MHz signal that is supplied to the microscope stage is attenuated by a known amount with precision rf attenuators (Kay Electronics models 1/435, 435, and 0-700B), thus reducing the brightness of the TV image to a point where the interference image just becomes indistinguishable from the background noise in the image, as determined visually by the operator. The specific value of inserted electrical attenuation is recorded, and the measurement is repeated without the specimen, in place, but at the same position in the sound field. Each of these attenuation determinations has an uncertainty of ±2 dB. The difference in these two attenuation values represents the insertion loss. To obtain the attenuation coefficient, the insertion loss is found for at least three different specimen thicknesses, and from the slope of the insertion loss vs specimen thickness, the ultrasonic attenuation coefficient is determined. This procedure is insensitive to constant reflective losses at tissue–saline interfaces in the attenuation coefficient value. The thickness of the specimens used for these determinations ranged from 0.7 to 2.44 mm, with a maximum uncertainty of 15 μm. The attenuation coefficient determined with the SLAM has an estimated uncertainty of 2 Np/cm.

RESULTS AND DISCUSSION

The attenuation in bovine liver was measured at 1.37, 4.15, 6.90, and 9.65 MHz using the radiation force balance, and at 100 MHz using the scanning laser acoustic microscope. The results are summarized in Table I and represent data from eleven liver specimens. It should be noted that in many cases, samples from the same liver specimen were measured at a minimum of three of these frequencies.

A least squares fit of the attenuation coefficient (Np/cm) vs frequency (in MHz) to the 1–10 MHz data (based on a total of 108 measurements) yields

$$A = 0.043\, f^{**}1.266 \quad (r^{**}2 = 0.895) \qquad (2)$$

and to the 1–100 MHz data (based on 116 measurements) yields

$$A = 0.043\, f^{**}1.270 \quad (r^{**}2 = 0.954), \qquad (3)$$

where A is the attenuation coefficient, and $r^{**}2$ is the best fit parameter (unity is a perfect fit). Equation (3) is plotted in Fig. 2, along with regions which include the range of other available measurements of attenuation coefficients (180 measurements) and absorption coefficients (4 measurements) in fresh, nonpathologic liver tissue at frequencies above 1.0 MHz.[1,2] About one-half of the attenuation and all of the absorption measurements were in bovine liver tissue. The region which includes all the attenuation measurements would have been basically the same size and shape if only the bovine liver tissue measurements had been included.

Several observations can be made concerning these results:

(1) Previous fresh liver attenuation coefficient data[1,2] have been in the 0.082–14 MHz range. Hence, the measurements reported herein represent a significant increase in the frequency range of liver attenuation coefficient mea-

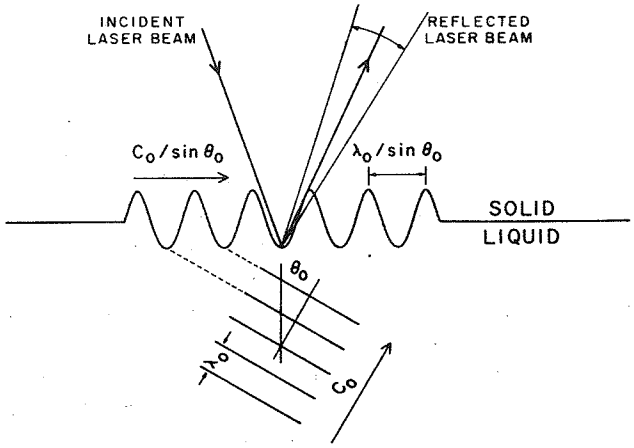

FIG. 1. Schematic representation of the dynamic ripple diffraction technique employed by the Scanning Laser Acoustic Microscope.[6]

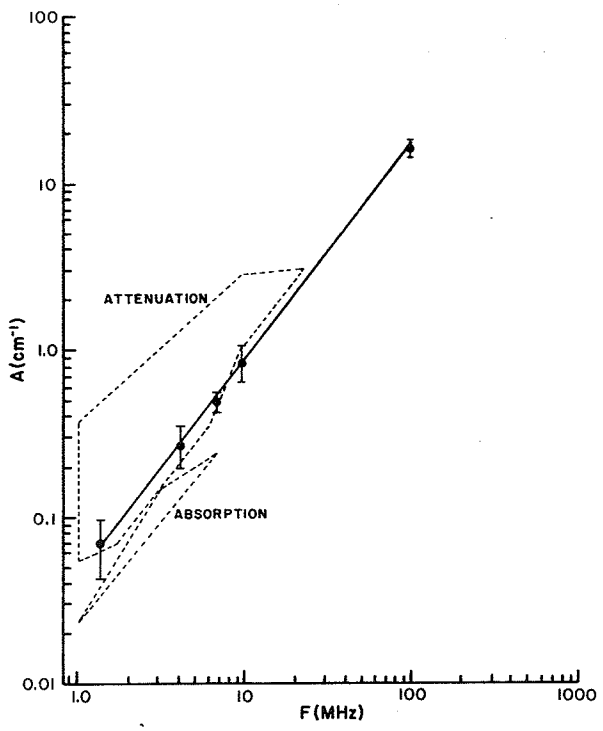

FIG. 2. Plot of Eq. (3), and ranges of previous measurements of attenuation and absorption coefficients in fresh liver.

surements. Of even greater importance is the fact that in many cases the same specimen was measured at several frequencies. This, in fact, represents the greatest frequency range over which the attenuation coefficient has been determined for the same specimen.

(2) A comparison of the exponents in Eqs. (2) and (3) (1.266 and 1.270, respectively) shows that the frequency dependence of the attenuation coefficient in bovine liver is essentially constant over the entire 1–100 MHz range. This is the first time that such a result has been shown. This is a particularly interesting result because it had been speculated that as the frequency was increased, the attenuation would be dominated by the attenuation due to the water in the tissue, and would approach a frequency dependence of f**2 (the frequency dependence of water). In partial support of this notion, it should be noted that Kessler's[8] measurements of the attenuation coefficient of mouse kidney at frequencies of 96 and 222 MHz (also measured on the SLAM) suggested that a square law or greater dependency upon frequency may exist at frequencies greater than 100 MHz.

(3) The frequency dependencies reported here, namely 1.266 and 1.270, are slightly greater than has been observed from various literature compilations. For instance, Goss[9] reported that literature values for the attenuation coefficient in liver showed a frequency dependence of 1.13, and an analysis of each of the liver studies reported in the compilation by Goss *et al.*[1,2] has shown that all of them exhibit frequency dependencies between 0.48 and 1.15.

(4) A comparison was made between the magnitude of the attenuation coefficient as given by Eq. (3), and those published for fresh liver in the compilations by Goss *et al.*[1,2] With very few exceptions, the values given by Eq. (3) were lower than the published values (Fig. 2). This may be due in part to the fact that the measurement methods employed here are insensitive to phase cancellation artifacts. This has not always been the case with previously reported results.

(5) Published data on absorption in liver[1,2] were used to calculate the ratio of attenuation to absorption. The results ranged from 1.3 to 2.2 (Fig. 2). This is also slightly lower than previously reported results, and is consistent with the lower attenuation.

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