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ULTRASONIC ATTENUATION AND ABSORPTION IN LIVER TISSUE

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Abstract—A large range of values for ultrasonic attenuation and absorption coefficients of tissues are reported in the literature. An important distinction both practically and theoretically is the magnitude of the true absorption, which characterizes the rate of conversion of ultrasonic to thermal energy, as compared with the total attenuation of the ultrasonic signal as it propagates through tissue.

The magnitudes of these quantities were studied in bovine liver. Total attenuation was measured, in the range of 1–6 MHz, by both phase sensitive and phase insensitive insertion loss techniques. Ultrasonic absorption was determined by two thermal techniques. The standard "transient thermoelectric" or rate-of-heating method, and a new measurement technique based on the temperature decay following a short ultrasonic pulse were employed for the determination of the ultrasonic absorption coefficient.

The results demonstrate that the ultrasonic amplitude attenuation and absorption coefficients at low megahertz frequencies are not significantly different in liver. The mean values cluster around 0.05 nepers/cm/MHz (0.4 dB/cm/MHz). The sample-to-sample variation is indicated by the standard deviation in the measurements of 0.01 nepers/cm/MHz (0.09 dB/cm/MHz) or less.

The results show that in liver tissue, absorption is the dominant feature of attenuation over this frequency range.

Key Words: Ultrasound, Attenuation, Absorption, Tissue characterization.

INTRODUCTION

The measurement of ultrasonic absorption and attenuation in biological tissue is relevant to practical uses of diagnostic and therapeutic ultrasound. Theoretical understanding of the underlying mechanisms also requires careful measurement on tissues of known composition and structure. Although these data are essential for medical ultrasound applications, a wide discrepancy exists in the reported values. Values as low as 0.02 nepers/cm/MHz have been measured for the amplitude absorption coefficient of liver (Goss *et al.*, 1979); and values as high as 0.14 nepers/cm/MHz are reported for the amplitude attenuation coefficient of liver (Goss *et al.*, 1975).

It has been suggested that the difference between attenuation (A) and absorption (α) for a tissue yields information which can be used in tissue classification as well as in understanding the basic mechanisms involved (Polhammer and O'Brien, 1981).

Absorption mechanisms converts the energy of an acoustic wave to heat as the wave propagates through a medium. A plane, ultrasonic wave in an absorbing medium will lose intensity as

$$I(x) = I_0 \, e^{-2\alpha x} \tag{1}$$

where α is the amplitude absorption coefficient of the medium, x represents the distance traveled and the I_0 is the intensity at $x = 0$. At each point in the absorbing medium, heat is generated at a rate of $2\alpha I(x)$ W/cm^3. This volumetric heat generation raises the temperature of the sample and embedded thermocouple techniques can therefore be used to measure the absorption coefficient.

The total attenuation of an ultrasonic wave as it passes through a sample of tissue is caused by absorption combined with the effects of scattering or internal reflections which may divert the acoustic wave from the forward sound path without significant local conversion of the acoustic energy to heat. If the sources of these effects are distributed uniformly throughout the tissues, then both absorption and scattering are homogeneous, bulk properties of the tissue. One could define an amplitude attenuation coefficient A (where $A > \alpha$) such that the intensity loss is given by:

$$I(x) = I_0 \, e^{-2Ax} \tag{2}$$

which could be measured by appropriate insertion loss techniques.

Surface reflection artifacts could influence attenuation measurements but usually these effects are small for soft tissues in saline and are comparatively easy to determine and eliminate from the data.

Large piezoelectric receivers are notoriously sensitive to phase-cancellation errors in insertion loss measurements (Marcus and Carstensen, 1975; Busse *et al.*, 1975). In extreme cases, it can make nearly lossless materials appear to have extremely large attenuation coefficients. Although this phenomenon may be important in the operation of certain devices, it should not be considered a property of the tissues but rather

an artifact of the measurement method. There is little merit in defining a phase sensitive as opposed to a phase insensitive attenuation because if the two measurements differ the phase sensitive values are prone to be highly variable and dependent upon orientation of the receiver and the tissue sample.

The results of this study show that when care is taken to eliminate artifacts of measurement, most of the techniques which have been used to measure attenuation and absorption, when applied to soft tissue such as bovine liver, given substantially the same magnitudes.

MATERIALS AND METHODS

Sample preparation. For these experiments, bovine liver sections were procured immediately after slaughter and placed in chilled, degassed, physiological saline. Some of the samples were stored at 4°C for as long as 4 hr before measurement. Other sections were frozen and later thawed for the tests. Ultrasonic properties of fresh and properly handled frozen-thawed liver samples have been shown to be similar. Pressurization of thawed samples is a very effective means of eliminating entrapped gas bubbles (Frizzell *et al.*, 1979). If gas (i.e. nitrogen) is used to pressurize a chamber containing the liver sample, some form of barrier is required between the gas and the sample with its surrounding fluid. In the absence of such a barrier, the gas is driven into the tissue sample, and will reappear as a multitude of small bubbles shortly after decompression. For these studies, a sealable container of soft polyethylene was used to contain the tissue and saline within the pressure chamber. The plastic container transmits hydrostatic pressure, while preventing pressurized nitrogen from dissolving into solution. Thawed samples were held at 400 psi for 30 min.

For all measurements, tissue slices approx. 2 cm in thickness were held between plastic film acoustic windows which by themselves had negligible attenuation. The sample faces were held parallel at a 2 cm ($\pm$ 1 mm) thickness by the retaining membranes. All measurements were taken at 20°C.

Total acoustic power insertion loss measurements. To obtain a phase-insensitive attenuation measurement a radiation force receiver was used. A natural rubber absorbing target, connected to an electronic balance, intercepted the entire ultrasound beam of a 1.2 cm dia. unfocused transducer operating at 1.2, 3.4 or 5.6 MHz continuous wave operation. Source-receiver separation was 7 cm. The change in acoustic power resulting from insertion of a tissue sample can be attributed to the attenuation mechanisms within the material.

Piezoelectric receiver method. For a phase-sensitive attenuation measurement a 1.2-cm-dia. plane piezoceramic receiver was used. In this technique, the source and receiver were held at a fixed separation of 8 cm and aligned acoustically. A narrow band, 75 μsec pulse was used with carrier frequencies of 3.0, 5.5 and 7.0 MHz. Since phase cancellation, beam diffraction, and refraction effects in all cases would lead to increased apparent attenuation, the tissue sample was inserted between the transducers and painstakingly oriented until the maximum received signal was obtained. These insertion loss values were used in calculation of the attenuation coefficient. Unless those precautions were taken, it was common to obtain attenuations which were greater by a factor of two or more.

Pulse-echo measurements. Rapid acquisition of tissue attenuation data can be achieved using a broad band pulse in a pulse-echo mode of operation (Lele and Namery, 1974; Kak and Dines, 1978; Bamber *et al.*, 1977). A reference spectrum is obtained by using an acoustic reflector carefully aligned to be perpendicular to the beam axis. A tissue sample is then inserted between the transducer and the reflector and a new spectrum is obtained of the signal after a round-trip through the material. The difference of the two spectra yields twice the attenuation of the sample. If a broad band pulse is used, the attenuation over a range of frequencies is obtained. Since the receiver is piezoelectric, this measurement is sensitive to "phase cancellation errors" (Marcus and Carstensen, 1975; Busse *et al.*, 1975) which create large measurement artifacts from system mis-alignment or inhomogeneities within the sample medium and lead to unreasonably large "attenuation coefficients." These reflect measurement artifacts, however, and not inherent properties of the tissue. If the reflector is properly aligned before introduction of the sample, these artifacts will act to increase the apparent attenuation. For this reason *minimum* values of attenuation obtained through careful alignment of the sample are reported here.

For these studies, a 1.2-cm dia. broad-band transducer was used to transmit acoustic pulses. A thick block of solid aluminum was used as the acoustic reflector. The transducer, sample, and reflector were carefully aligned to obtain the maximum signal. The usable bandwidth of the acoustic signal was 2–5 MHz as measured by a Tektronics Spectrum Analyzer. The magnitude of the signal's frequency spectrum was recorded at 0.5 MHz intervals between 2 and 5 MHz, with and without the presence of the liver sections. The attenuation values were then obtained from a plot of total attenuation vs frequency.

Transient thermoelectric, rate-of-heating method. True absorption mechanisms convert ultrasonic en-

ergy to heat. Accordingly, by exposing a sample to ultrasound at known intensity, and measuring the resulting temperature rise, the absorption coefficient can be determined. The governing equation is

$$\rho C \, dT/dt = 2\alpha I \qquad (3)$$

where ρ and C are the tissue density and specific heat, T is the temperature and t is time.

An embedded thermocouple is used to measure temperature elevation. Measurement errors result initially from viscous heating around the thermocouple-tissue interface and later by conduction of heat away from the region heated by the ultrasonic beam (Fry and Fry, 1954a,b; Goss *et al.*, 1977). For this study, 50 μm dia. copper-constantan thermocouples were placed 2 mm deep in liver samples to record temperature elevations. The rate-of-heating was measured 0.5 seconds after the beginning of irradiation to minimize viscous heating and heat conduction errors (Goss *et al.*, 1977). The product ρC for the tissues was taken to be 3.8 J/cm^3 (Bowman *et al.*, 1975). Acoustic intensities were measured for each frequency with a 0.16-cm dia., spherical, steel radiometer (Dunn *et al.*, 1977). A 1.2-cm dia. transducer was used to provide the ultrasound at 1.2, 3.4 and 5.6 MHz. At each frequency, samples were centered on the beam axis by observation of temperature increments with short pulses of sound. Intensities used in the calculation of absorption coefficients were corrected for absorption in the 2 mm of liver tissue above the thermocouple.

Thermal pulse-decay method. As mentioned above, direct thermocouple measurement of the temperature elevation caused by ultrasound is clouded by the viscous heating effect around the metal wire. The rate-of-heating method minimizes this artifact by using a long ultrasonic "on time". Since the viscous heating effect is restricted to a small volume at the thermocouple-tissue interface, this excess heating has a short thermal time constant. Thus, the viscous heating effect approaches equilibrium rapidly, and by waiting 0.5 sec or so after the start of insonation, its effect on the rate of heating measurement is minimized (Goss *et al.*, 1977).

A different approach is taken by the thermal pulse decay method. A short (less than 0.1 sec) pulse of ultrasound is used to heat a region of tissue. An embedded thermocouple registers the true heating caused by absorption mechanisms, plus the effect of viscous heating around the wire. After the pulse, heat flows by conduction to the surrounding, cooler regions. As before, since the viscous heating effect occurs in a small volume surrounding the thermocouple, it displays a very short thermal time constant. Figure 1 shows that within two seconds following an ultrasonic pulse in liver tissue, the viscous heating effects have become negligible, and the thermocouple thereafter records the decay of heat which originated from true absorption of ultrasonic energy.

By ignoring the thermocouple response immediately after the pulse, and by explicitly accounting for heat conduction away from the directly heated re-

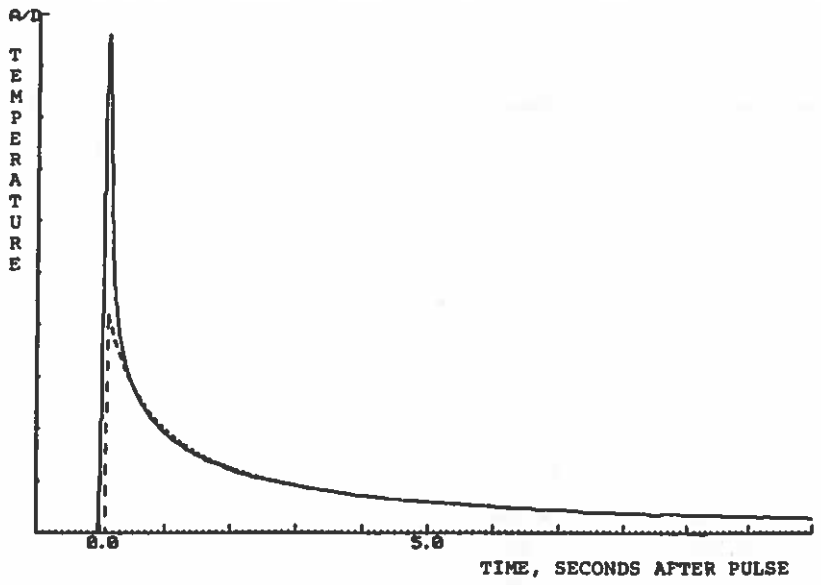

Fig. 1. The pulse-decay technique. This shows the temperature decay in tissue following an ultrasonic pulse. ———, Measured thermal history; - - - -, theoretical "retro-fit" neglecting viscous heating effects. From the theoretical fit the absorption coefficient is determined.

gions, a measurement of tissue heating can be obtained which is free of the artifact of localized viscous heating (Parker, 1981). The information is used to calculate the absorption coefficient of tissue.

A detailed description of this technique, the mathematical models used and comparison with the rate-of-heating method are described in detail elsewhere (Parker, 1981; Parker and Lele, 1982). For the work reported herein, a focused beam at 2.25 MHz was used to heat a region of tissues. The liver absorption coefficient derived from the pulse-decay technique was found to be very close to those determined by the rate-of-heating method as noted previously (Parker, 1981).

Test comparisons. A rubber casting compound was used to construct a 2.15-cm-thick absorbing sample material with an embedded 50-μm-dia. thermocouple.

The attenuation and absorption values for this homogeneous medium were measured at 3.4 MHz using the acoustic power insertion loss method and rate-of-heating method, respectively. After five measurements of the sample at 20°C the amplitude attenuation values were found to be 0.153 ($\pm$0.003) nepers/cm/Mhz, and the amplitude absorption coefficient was 0.161 ($\pm$0.015) nepers/cm/MHz. The absolute values are in close agreement, as expected. The higher variation in absorption measurements can be traced to variations in the measurement of absolute intensity via the steel ball radiometer, the uncertainty in beam alignment with respect to the thermojunction, and the errors in determining the rate of heating from a strip chart recording.

This test comparison confirms that the experimental techniques are capable of measuring equal attenuation and absorption coefficients of a homogeneous, absorbing medium.

RESULTS AND DISCUSSION

Ten samples' properties were measured using the radiation force method (attenuation) and the rate-of-heating method (absorption). The data are shown in Fig. 2. Standard deviations representing sample-to-sample variability are shown by vertical bars in the figure. Lines intercepting the mean values at three measured frequencies are drawn for clarity. It is tempting to conclude from the lines that attenuation is demonstrably greater than absorption over all measured frequencies. This conclusion is not supported by the statistics of the measurements, however. A student's t-test for paired data applied to the measurements showed no statistically significant ($p > 0.05$) difference between attenuation and absorption values at 1.1 or 3.4 MHz. The only statistically significant difference between measurements of attenuation (by the radiation force method) and absorption (by the rate-of-heating method) for the 10 bovine liver samples occurred at 5.6 MHz, where absorption comprises 82% of the total attenuation. The small but positive slope of the attenuation curve suggests the well known frequency dependence, increasing as f^n where $1.0 < n < 1.2$ (Goss *et al.*, 1975).

All five measurement techniques discussed in "Materials and Methods" were used to collect data on three liver samples (a subset of the ten samples

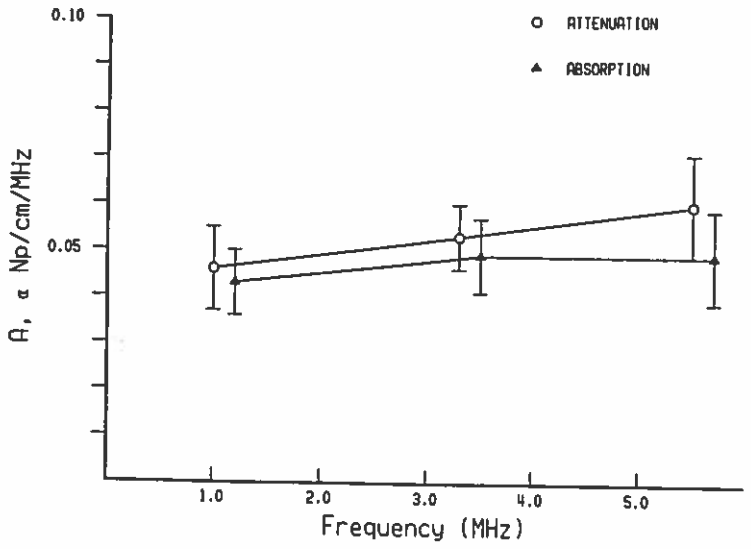

Fig. 2. Absorption and attenuation vs frequency in ten bovine liver samples. Absorption is measured by the rate-of-heating technique, attenuation by the radiation force method.

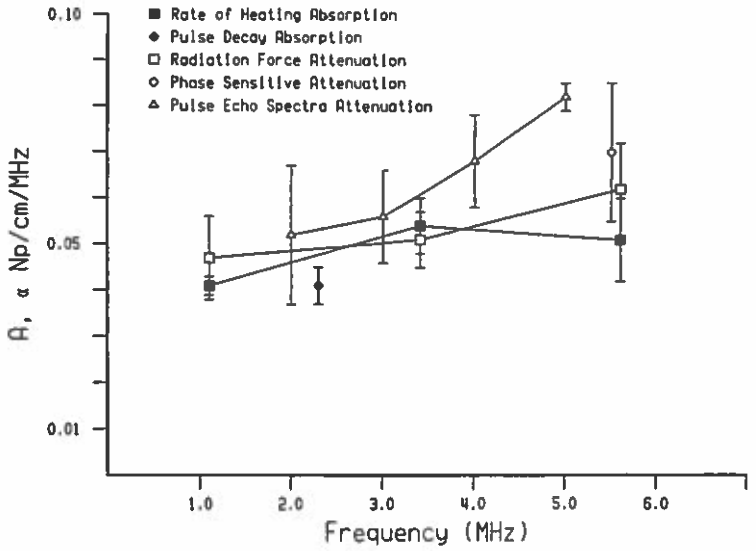

Fig. 3. Absorption and attenuation vs frequency for three liver samples. Absorption is measured by rate-of-heating and pulse-decay techniques. Attenuation is measured by the radiation force method, as well as by the phase-sensitive insertion loss and pulse-echo spectra techniques.

represented in Fig. 2). These results are shown in Fig. 3. Data obtained using the pulse-decay method for measuring absorption agree with results from the rate-of-heating technique. Although the attenuation measurements include "phase sensitive" techniques, a close matching of attenuation and absorption measurements is obtained below 3 MHz.

An interesting feature of these curves is the relative increase in phase-sensitive attenuation measurements above 3 MHz. Much higher values could have been obtained using these methods, if the precautions described in the Materials and Methods section had not been strictly followed.

An important distinction when considering "phase-sensitive" techniques is the difference between the effects of a finite aperture phase-sensitive receiver, and the effects of local tissue inhomogeneities which distort wavefronts. The distinction is important in eliminating artifact from experimental measurements of attenuation. The minimization of errors caused by the use of a finite aperture, phase-sensitive receiver can sometimes be accomplished by simply reducing the aperture size (Busse *et al.*, 1975; Busse and Miller, 1981). In attenuation measurements, the effects of tissue as an "aberrating" medium must still be considered even when the detector is small and energy sensitive. Wavefront distortion will occur as the ultrasonic signal propagates through many wavelengths of inhomogeneous material. This effect will add an element of variability to measurements.

To illustrate this effect, a small hydrophone was used to measure the underwater acoustic field of a narrow band 3.4 MHz, 1.25 cm dia. transducer, with and without an intervening sample of liver. The signal pulse length was 70 μsec, and values of received pressure were read from an oscilloscope. A hydrophone probe was moved in 1-mm increments in the radial direction to obtain a scan at 8 cm range (near the transition zone of the field). The measured water-only beam pattern is shown as the solid line in Fig. 4. Next, a 2.2-cm-thick sample was carefully positioned normal to the beam, at a range of 1 cm from the face of the hydrophone. The source strength was increased 4 dB to compensate for the effects of attenuation (determined from radiation force measurements). If the material effects were limited to attenuation, the scans would overlap. Instead, an altered beam pattern was observed which changed with shifts in the lateral position of the liver. Two representative measured beam patterns are shown in Fig. 4. The general effect of the liver samples is to distort the original beam pattern. Significantly, the sharp peak and null patterns of the undisturbed beam are smeared. This is not surprising since the peak and null patterns are caused by interference of phase fronts from the source transducer. Local variations in acoustic properties such as phase speed or attenuation, caused by inhomogeneities within the tissue, will perturb this summing of phase fronts. Beam distortion is known to affect imaging of tissues (Banjavik *et al.*, 1978; Halliwell, 1978).

The implications for attenuation measurements are clear. In this case the use of a small probe (phase sensitive or insensitive) to measure peak or on-axis

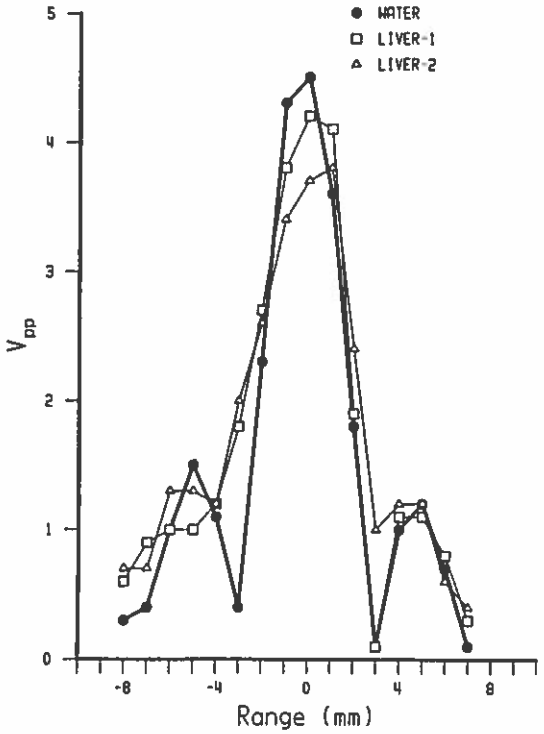

Fig. 4. Measured beam patterns from an unfocused 3.4 MHz source. ———, Water-only path; ———, liver sections inserted between source and scan plane, with source strength increased 4 dB to compensate for attenuation effects. These demonstrate the nature of liver as an "abberating" medium.

intensity transmitted through the liver would give variable results. This measurement approach would generally indicate higher attenuation values than those derived from radiation force measurements. The forward wave energy is not only reduced by the liver, but also redistributed. A single point measurement cannot describe the overall situation. In contrast, the output from a large phase-insensitive receiver which integrates over the entire insonified region is not affected by small redistributions of energy, and therefore gives more consistent results showing lower attenuation. If a small hydrophone is used for attenuation measurements, multiple points should be used to reconstruct beam patterns over a plane. This enables a summation to be made of the total power in the transmitted wave, before and after insertion of an aberrating medium. In short, the use of a small aperture piezoelectric probe minimizes artifacts caused by the phase sensitive nature of the receiver, but "single point" transmission measurements are inadequate where beam distortion occurs.

SUMMARY

In the ultrasonic literature the highest values of the attenuation are nearly seven times greater than the lowest values which have been reported for the absorption of soft tissues such as liver. To ascertain what part of that difference is experimental error and what is a true difference between attenuation and absorption, we used a variety of measurement techniques on samples of bovine liver tissue. One technique which was not represented in our study was the measurement of attenuation using an acoustoelectric receiver. However, we have no reason to believe that the results would differ from the phase insensitive radiation force measurements.

Our purpose was not to find the best measuring technique, but rather to determine the difference between attenuation and absorption for a representative tissue. The values for amplitude attenuation and absorption cluster around 0.05 nepers/cm/MHz, with sample to sample variations in the measured values on the order of ± 0.01 nepers/cm/MHz. The attenuation and absorption values (summarized in Fig. 2) showed no statistically significant difference at 1.1 or 3.4 MHz. A statistically significant difference between absorption and attenuation was found at 5.6 MHz, where the mean value of absorption is 82% of the mean attenuation coefficient.

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