Reprinted from *Ultrasonic Tissue Characterization II*, M. Linzer, ed., National Bureau of Standards, Spec. Publ. 525 (U.S. Government Printing Office, Washington, D.C., 1979).

ELEMENTS OF TISSUE CHARACTERIZATION

Part II. Ultrasonic Propagation Parameter Measurements

S. A. Goss, R. L. Johnston, V. Maynard, L. Nider,
L. A. Frizzell, W. D. O'Brien, Jr., and F. Dunn

Bioacoustics Research Laboratory
University of Illinois
Urbana, Illinois 61801, U.S.A.

Methods employed at the Bioacoustics Research Laboratory of the University of Illinois for the determination of ultrasonic propagation properties of biological media are described, with attention devoted to attenuation, absorption and velocity measurements of both longitudinal and shear ultrasonic waves. These include systems specifically for the ultrasonic characterization of soft tissues and for solutions of biologically significant macromolecules. Each method is presented in terms of theory, limitations, applicability, and possible sources of error. Important new techniques from other laboratories are also discussed.

Key words: Ultrasonic absorption; ultrasonic attenuation; ultrasonic instrumentation; ultrasonic measurements; ultrasonic spectroscopy; ultrasonic tissue characterization; ultrasonic tissue parameters; ultrasonic tissue signature; ultrasonic velocity.

1. Introduction

A number of measuring techniques have been developed, or are otherwise employed, at the Bioacoustics Research Laboratory of the University of Illinois in the research efforts associated with the propagation characteristics of ultrasound in biological tissue, and toward a basic understanding of the mechanism(s) responsible for the acoustic properties exhibited by those tissues. These include specialized systems for biological liquids and for soft tissues which are capable of measuring attenuation, absorption, and velocity of longitudinal ultrasonic waves as functions of frequency, temperature and, where appropriate, pH and ambient pressure. In addition, a system is presently being developed to measure the shear acoustical properties of biological materials as a function of frequency and temperature. These techniques, thirteen in all, are described in this paper in terms of parameters measured, theory, application, precision/accuracy, frequency, and temperature range, so that each technique may be assessed separately.

2. Attenuation and Velocity Measurements in Soft Tissue

Attenuation of longitudinal waves in soft tissue specimens are determined using the radiation force, pulse transmission, and standing wave techniques. Velocity measurements in soft tissues can be made by observing the time of flight of an acoustic pulse through a known path length of a tissue sample, and by acoustic interferometry.

A. Attenuation: Radiation Force Method

The phenomenon of radiation force provides a primary method for the measurement of the second order quantities of intensity and power [1-4][1]. Radiation force is a direct result of energy transport by the sound wave, and is equal to the time rate of change of momentum of the wave. Thus a continuous wave incident on a reflecting or absorbing object will produce a time independent force on that object equal to, and in the direction of, the time rate of change of momentum. Consequently, a sensitive balance can be employed to measure the radiation force exerted by an ultrasonic beam incident on a suspended target. It then follows that force, measured as a change in tension of the suspending wire, on a perfectly absorbing target intercepting a vertically directed sound beam is

$$F = \frac{P}{c} \qquad (1)$$

where F is the measured change in tension of the suspending wire, P is the time average power intercepted by the target, and c is the velocity of propagation. Thus, in water, 1 milliwatt of incident power will exert a force equivalent to a weight of 67 μg, which is measured as an apparent target weight change. Similarly, a perfectly reflecting target intercepting a vertically directed sound beam will produce a measurable change in suspension tension, given as

[1]Figures in brackets indicate literature references at the end of this paper.

$$F = \frac{2P}{c} \cos^2\theta \qquad (2)$$

where θ is the angle between the beam axis and the normal to the target surface. Consequently in water, when the target is inclined 45° to the vertical, the sensitivity of the balance is again 67 μg/mW of intercepted power. Thus, a phase insensitive, frequency independent technique for power and intensity attenuation measurements becomes available by interposing a tissue sample between the sound source and the sound intercepting target, by determining that power lost or redirected with the tissue sample in place compared with measured power with the tissue absent [2,5,6]. Care must be taken to insure that the system's sensitivity is actually that that predicted for such idealized cases. The attenuation present in the liquid media (water or physiological saline) will inevitably cause acoustic streaming which will lead to systematic errors in the power estimation. The streaming effect can be minimized by positioning an acoustically transparent barrier of, for example, stretched polyethylene, directly in front of the target.

A second possible source of error is due to the inherent insensitivity of the measurement system to horizontal forces. For maximum accuracy, the main lobe of the sound beam should be directed precisely vertically. Because the beam may contain components of momentum in the plane perpendicular to the main lobe, the radiation force balance method does not respond to all the energy emitted by the source. As the temperature of the target increases due to the fact that energy is being deposited in it, thermal expansion of the target can introduce an error by virtue of increasing its buoyancy and resulting in an apparent target weight change. The ultimate accuracy of the radiation force balance method is limited by noise associated with mechanical vibrations and atmospheric perturbations, which can be reduced by proper design considerations. However, in the presence of pressure gradients, the target is acted on by forces proportional to the target volume, making a small target advantageous.

B. Attenuation: Pulse Transmission Method

The pulse transmission technique is applicable to measurements of sound pressure attenuation in various types of biological specimens [7,8]. The instrumentation consists of a transmitting ultrasonic transducer immersed in a bath (usually physiological saline) which serves as the acoustic coupling medium and aligned axially with either a reflecting surface or a receiving transducer. The associated electronic instrumentation provides an RF pulse to the transmitting crystal and also triggers the sweep of an oscilloscope a variable, but selected, length of time after the initiation of the transmitting pulse. An oscilloscope serves to display the amplified received signal. Measurement of sound attenuation can be performed as follows. First, with no sample in the ultrasonic path, the gain of the receiving amplifier is adjusted to give a signal display of a predetermined amplitude. Then, with the specimen in place, the gain of the system is adjusted, either by increasing the receiving amplifier gain or by removing electrical attenuation in the signal path, to compensate for the loss of acoustic energy in the sample to bring the display trace to the same preselected amplitude. The attenuation measurement is repeated for other samples of the same material, but of varying thicknesses. The measured attenuation values are then plotted versus sample thickness, and the slope of the resulting straight line yields the attenuation per unit length of the specimen.

C. Velocity: Pulse Transit Time Method

Velocity measurements in tissue can be made simply by measuring the time of flight of an acoustic pulse over a known path length of the specimen [9,10]. Differential techniques should be used to minimize uncertainties in path length and time delay measurements, such as the uncertainty of the actual location of the active face of the transducer. If commercial transducers are employed, additional allowances must be made for the quarter-wave impedance matching layer on their surface. Errors may be encountered in time of flight measurements of the acoustic signal in inhomogeneous media, such as tissues. Since a short duration acoustic pulse contains a broad spectrum of components, the frequency dependent effects of velocity dispersion, attenuation, and multiple phase shifts at tissue interfaces, can distort and delay the signal [11]. Techniques which rely on the detection of the leading edge of the received acoustic pulse for timing are prone to the greatest inaccuracies under such distortions.

D. Velocity: Acoustic Interferometric Method

Nearly an order of magnitude improvement in accuracy can be realized by the use of acoustic interferometric techniques employing CW excitation [12,13]. By sandwiching the tissue sample between transmitting and receiving transducers, or between a transducer and a reflecting target, a standing wave is created in the tissue sample. It has been shown that tissue may be distorted in this manner up to 25 percent without affecting the measured velocity [12]. By monitoring the maxima and minima of the received acoustic signal as the path length is changed, the standing wave pattern and the wavelength of sound in the tissue can be measured. This information combined with the frequency employed for the measurement yields the velocity of sound in the sample. Alternatively, the path length may be held fixed and the frequency of excitation changed. The change in frequency necessary to move from one mode in the standing-wave pattern to the next is directly proportional to velocity. The accuracy of such interferometric techniques, under tight temperature control, can reach about 0.1 percent.

E. Attenuation and Velocity: Standing Wave Method

While the measuring techniques described above may be applied to the measurement of the ultrasonic absorption and velocity of most biological tissues, some specimens require specialized techniques.

Lung, by virtue of its extraordinarily high attenuation, presents some problems in determination of its ultrasonic propagation properties. The following method has been developed and yields results in agreement with other similar ones [8].

Briefly, the lung is suspended in the sound field between the sound source and an absorption chamber, to eliminate standing waves beyond the lung. The transient thermoelectric probe [14] provides a convenient acoustic detector to investigate (A) the acoustic field between the specimen and the source, to determine the axial standing wave pattern, and (B) the acoustic field between the specimen and the absorption chamber, to determine the wave amplitude transmitted beyond the lung specimen 15 (see fig. 1). Probing the field in A yields the fraction of incident energy reflected at the lung-saline interface. Assuming that infinitesimal wave acoustics prevails and that the attenuation in the lung is sufficiently great that multiple reflections within the specimen need not be considered, the speed of sound in the lung can be obtained from

$$(\rho v)_{lung} = (\rho v)_{saline}\left(\frac{1}{SWR}\right) \qquad (3)$$

where the ρ's are the known densities and SWR is the observed standing wave ratio (between the source and specimen) and is greater than unity. The attenuation coefficient per unit path length is determined from a knowledge of the energy reflected at the two lung-saline interfaces, the thickness of the sample, and the acoustic intensity detected by the probe in B in accordance with the relation

$$I_1 = I_0 e^{-2\alpha l} \qquad (4)$$

where I_0 and I_1 are, respectively, the acoustic intensities at the lung-saline interface nearest to and farthest from the source, l is the thickness of the sample, and α is the attenuation coefficient.

3. Absorption Measurements in Soft Tissue

The transient thermoelectric method is well suited for determining acoustic absorption in small volumes of highly absorbing liquid or tissue *in vivo* as well as *in vitro*, and may be the only method, applicable to tissues, that measures directly absorption rather than attenuation. As such it is invaluable in determining the portion of attenuation due to absorption *vis-a-vis* that portion due to scattering effects [16,17]. The method requires that a thermocouple junction of small diameter relative to the wavelength be implanted in the sample. The thermocouple and surrounding sample are then exposed to a plane traveling wave ultrasonic field having a temporally rectangular envelope of known intensity. The typical thermoelectric emf response to a 1 second ultrasonic pulse has an initial fast rise which results from conversion of acoustic energy into heat by the viscous forces acting between the thermocouple wire and the sample. This portion of the response approaches equilibrium very quickly with a magnitude that depends mainly upon the thermocouple wire radius, the viscous properties of the sample medium, the sound intensity, and the frequency. The fast rise is followed by a relatively linear rise (in the absence of thermal conduction processes) that is a result of absorption of the ultrasound in the surrounding medium. From a determination of the slope of the *linear* portion of the thermoelectric emf response as a function of time, the absorption coefficient can be calculated using the following relation,

$$\alpha = \frac{\rho C_p K}{2I}\left(\frac{dT}{dt}\right)_0 \qquad (5)$$

where α is the absorption coefficient (cm^{-1}), ρ is the density (g/cm^3), C_p is the specific heat at constant pressure (cal/°C g), K is the mechanical equivalent of heat (4.18 J/cal), I is the acoustic intensity (W/cm^2), and $(dT/dt)_0$ is the initial time rate of change of temperature due to absorption in the medium (°C/s). Effects limiting the

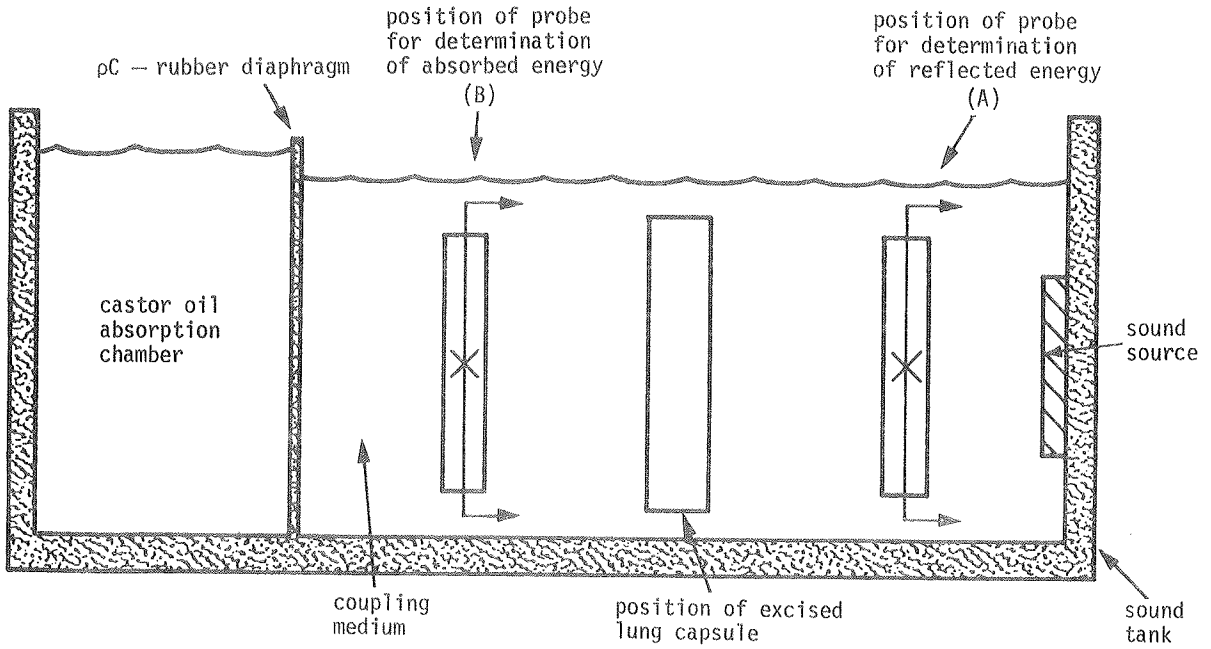

Fig. 1. Schematic diagram of the experimental arrangement for attenuation and velocity measurements in lung using the standing wave method [15].

accuracy of the method include any changes of the thermal properties of the tissue and/or the absorption itself with temperature. Provided the temperature rise during measurement is on the order of one degree or less these effects will be minimal. Dunn [18] has indicated that the total uncertainty in the determination of $(dT/dt)_0$ is of the order of 5 to 10 percent. In application of the transient thermoelectric method the initial value of α is determined by using the ultrasonic intensity that would be present at the site of the junction were the sample absent. The measured depth of the thermocouple and the initial value of α are then used to calculate intensity at the site of the junction by correcting for absorption using an iterative method [18] until the value of α converges (note that if the value of α and/or the depth of the thermocouple are too large, the value of α will not converge).

This technique has been used at frequencies as low as 0.26 MHz [19] and as high as 7 MHz [20] in tissues, but modifications have been used to 2 GHz in fluid media [21]. The lower frequency limit is determined by the value of the absorption, since too low an α yields too low a temperature rise to be detected accurately without having to increase the exposing intensity to unacceptable levels. Also, at low frequencies and absorptions the initial viscous heating portion of the transient thermal emf may become a major portion of the signal, making the subtraction process a relatively erroneous one. At higher frequencies the limitations on accuracy of the method, as described for tissues, are imposed by (1) availability of broad plane wave ultrasonic fields, (2) the size of the thermocouple relative to the wavelength so that the field is affected by scattering from the wire and thermal conduction along it, and ultimately by (3) nonconvergence in the iterative method of depth correction due to high absorption. The latter two limiting effects have not yet been troublesome in the region of application. The method has been found to be very useful and easily applied to most tissues over the frequency range from 0.5 to 4.0 MHz.

4. Absorption and Velocity Measurement in Biological Liquids

The primary mechanism(s) for the absorption of sound in biological material seems to occur at the macromolecular level. Carstensen, Li, and Schwan [22] discovered that the acoustical properties of blood are determined largely by the protein content, and that the absorption coefficient is directly proportional to the protein concentration, whether in solution or within the cell. Pauly and Schwan [23] have demonstrated that nearly two-thirds of the ultrasonic absorption of beef liver lies at the macromolecular level, with the remaining one-third due to structural features of the tissue. Since the primary mechanism for the absorption of sound in tissue seems to occur at the macromolecular level, it has been instructive to investigate the acoustic properties of simpler systems of macromolecules, such as solutions of biopolymers, with the hope that such studies may provide details applicable to the more complex systems.

Three measurement systems have been used at the Bioacoustics Research Laboratory to study acoustic absorption and velocity of solutions and suspensions of biomacromolecules as functions of molecular characteristics, pH, temperature, concentration, and frequency to aid in the elucidation of the mechanisms of interaction between ultrasound and biological tissues.

A. Absorption: High Frequency Method (5 MHz to 200 MHz)

The high frequency system applies pulsed ultrasound to solutions for the purpose of measuring absorption and velocity [24-26]. Two advantages of pulsed ultrasound are that heating effects and standing waves are virtually eliminated. Heating is proportional to the mean power. Using a 10 μs pulse at a repetition frequency of 300 pps results in a signal which is present for only 0.3 percent of the time. Thus the average intensity is much less than the peak intensity, and produces a negligible temperature rise. Since the signal is on for such a small fraction of the time, standing waves are not produced. The basic technique was first described by Pellam and Galt in 1946 [27]. Two transducers, arranged coaxially face each other at opposite ends of a cylindrical tank. The receiving transducer has its position fixed while the sending transducer moves toward or away from it at a constant speed. Assuming the relation governing the process is

$$p(x) = p(o)\, e^{-\alpha x} \qquad (6)$$

where $p(x)$ is the wave pressure amplitude at the distance x from the transmitting transducer, $p(o)$ is the pressure amplitude at $x = o$ (i.e., at the face of the transmitting transducer), and α is the absorption coefficient, the natural logarithm of the ultrasonic pressure amplitude is proportional to the signal path length. The proportionality constant, i.e., the absorption coefficient, can be found by monitoring the pressure as a function of intertransducer distance. A block diagram of the system is shown in figure 2. A pulsed rf signal is amplified, passed through a variable

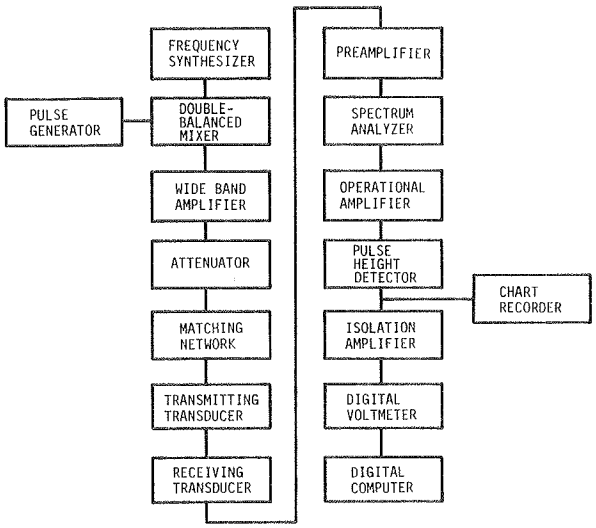

Fig. 2. Block diagram of instrumentation used for high frequency (5 to 200 MHz) ultrasonic absorption measurements in biological liquids.

attenuator, and impedance-matched to the sending transducer. Both transducers are air backed, gold-on-nickel plated X-cut quartz with a fundamental resonant frequency of, for example, 3 MHz, and operated at their odd harmonics. The wave passes through the liquid, where it is attenuated, and impinges upon the receiving transducer. Some of the ultrasound is then reconverted into an electric signal, while the remainder is reflected back through the liquid. A pulse height monitor continually detects the voltage peak of the primary signal (which is linearly proportional to p(x)) as the transducers are moved toward, or away from each other, and converts it to its natural logarithmic value. An on-line computer samples the output of the pulse height detector (a dc voltage proportional to the pulse height) via a digital voltmeter, calculates the attenuation coefficient, and corrects for diffraction effects by the method of Del Grosso [28] yielding the true absorption coefficient. Solute absorption is obtained by subtracting results of a solvent measurement from a solution measurement. Approximately one liter of sample is required for this technique, with measurement error in absorption of about 5 percent over the approximate frequency range 3 to 200 MHz. Greater errors occur in low absorbing liquids at low frequencies and are mainly due to relatively large diffraction effects.

B. Velocity: High Frequency Method

The acoustic velocity can be determined in the high frequency system to within 0.1 percent by measuring the period of the wave in the liquid. This is accomplished by superimposing a reference signal on the received signal, which results in an interference pattern. If the acoustic path length is changed at a constant rate, V_p, the period of this pattern is $T = (\lambda/V_p)$ where λ is the acoustic wavelength. Since $\lambda = c/f$, where f is the excitation frequency, the acoustic velocity is given as $c = TfV_p$. The period T is measured using a time interval counter to determine the time required for the interference signal to go through a greater number (100) of maxima.

C. Absorption: Low Frequency Method (0.3 MHz to 20 MHz)

In the frequency range of 0.3 to 20 MHz, a pulse technique developed by Schwan and Carstensen [29] is employed in order to avoid having to make unreasonable corrections for effects of diffraction phenomena. A sending and a receiving transducer face each other; the former is in the reference liquid and the latter in the test liquid (see fig. 3). An acoustic window separates the two liquids. The distance between the transducers remains constant while the entire transducer ensemble is moved horizontally at a constant speed from a position such that the acoustic path is primarily in the test liquid to a position such that it is primarily in the reference liquid. The reference liquid is chosen to be dispersionless and acoustically well characterized with a sound velocity as near to that of the unknown as possible. Water is an appropriate choice for dilute aqueous solutions. Since the path length is constant, diffraction effects, due to the inequality of the velocities in the two liquids, are corrected by a computer using the method of Del Grosso [28]. Liquid volumes of 1 to 4 liters are used in each chamber, depending upon frequency. As with the high frequency system, measurement error ranges from 2 to 10 percent. The pressure amplitude in the receiving transducer is given by

$$p_r = p_s e^{-\alpha_w(d-x)}\, e^{-\alpha_x x} \qquad (7)$$

where p_s is the pressure amplitude at the sending transducer, and the other symbols are described by figure 3. This equation can be rewritten as

$$\ln p_r = \ln p_s - \alpha_w d + (\alpha_w - \alpha_x)x \qquad (8)$$

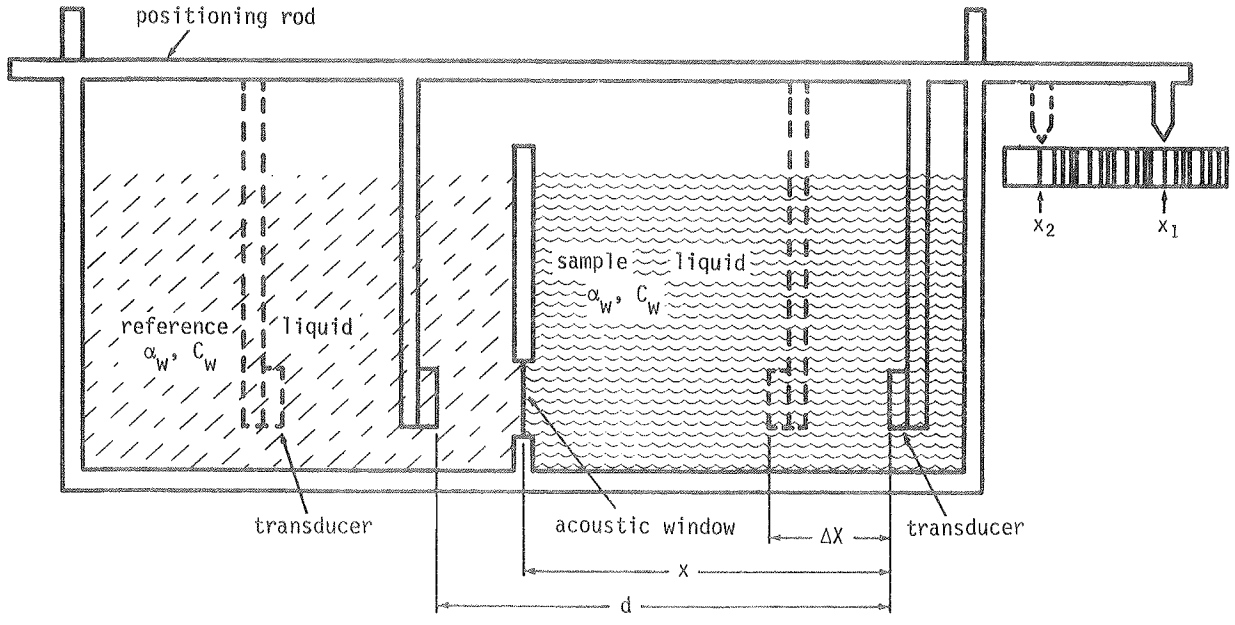

Figure 3. Schematic diagram of apparatus used for ultrasonic absorption and velocity measurements at low frequencies (0.3 MHz to 20 MHz) in biological liquids [24].

and α_x can be obtained graphically from the slope since α_w is known.

As the acoustic window (polyethylene) reflects a negligible, but constant amount of the acoustic energy, it will affect the results only by changing the intercept, i.e., the slope is not affected. Signal processing to obtain the desired absorption coefficient closely resembles that used in the high frequency system, described above [14,24].

D. Velocity: Low Frequency Method

Velocity of the acoustic wave can be determined with the low frequency system to 0.01 percent [30] by superimposing the received and reference signals in the same fashion as was done for the high frequency system. If the velocities in the test and reference liquids are different, moving the transducer assembly will change the acoustic path length. By observing the interference pattern on the oscilloscope, one can position the transducers such that a maximum occurs. If this is position 1 in figure 3, the number n, of acoustic wave lengths between the transducers is given by

$$n = \frac{d - x}{\lambda_w} + \frac{x}{\lambda_x} \qquad (9)$$

If the transducers are moved a distance, Δx, such that the interference pattern undergoes an integral number, m, of 2π phase shifts, the above expression becomes

$$n \pm m = \frac{d - x + \Delta x}{\lambda_w} + \frac{x - \Delta x}{\lambda_x} \qquad (10)$$

where the positive sign applies when the velocity in the test liquid is larger than that of the reference liquid, as is usually the case. Eliminating n between these equations, expressing wavelengths in terms of velocity and frequency, and a rearranging of terms yields

$$c_x = \frac{c_w}{1 \pm \frac{c_w m}{f \Delta x}} \qquad (11)$$

Note that some other procedure must be performed to determine the correct sign.

E. Absorption and Velocity Measurements in Small Liquid Volumes: Resonant Cavity Method

The various methods for determining the acoustic propagation properties of liquids discussed thus far were devised with little consideration for the volume of material necessary for measurement. The minimum volume required in the high frequency system is 500 ml, with greater volumes generally required at lower frequencies to avoid effects due to diffraction phenomena. The necessity for having such large volumes available becomes a serious problem when biological macromolecules are to be treated in solutions (usually in concentrations of 10 percent by weight), since only a few can be examined within the bounds of reasonable economics. It is possible to reduce the specimen volume to less than 50 ml using a resonance technique originally developed by Eggers [31], which employs two X-cut quartz disks, with a fundamental frequency of the order of 2 MHz, separated by a lucite ring, and forming a cylindrical cavity in which the specimen solution is placed. One transducer serves as a transmitter, and the other a receiver. When the transmitting transducer is excited with a sinusoidal voltage, standing waves in the specimen solution result at particular frequencies f_n that obey the resonance equation

$$\rho_q c_q \tan \frac{\pi f_n}{f_q} = \rho_s c_s \begin{bmatrix} -\tan \\ \cot an \end{bmatrix} \frac{\pi}{2} \frac{f_n}{f_s} \qquad (12)$$

where ρ_q and c_q are the density and speed of sound in the quartz transducers, and ρ_s and c_s are the corresponding quantities in the solution. The fundamental frequency of the transducer is f_q, and of the liquid-filled cavity, f_s. At the resonant frequencies f_n, the receiving transducer registers pronounced voltage peaks, whose frequency separation depends on the sound velocity of the specimen solution, and the half-power (3 dB) bandwidth Δf of which is related to the attenuation per wavelength, $\alpha\lambda$. Eggers [31,32] has shown that the velocity of sound v_f in an unknown medium may be related to the velocity of sound in some reference medium v_r by the expression

$$\frac{v_f}{v_r} = \frac{D_f}{D_r} \left| \frac{1 + 2(D_f Z_f - D_r Z_r)}{f_q Z_q} \right| \qquad (13)$$

where Z_f and Z_r are respectively, the acoustic impedance of the unknown and reference medium, and D_f and D_r are the respective separations in frequency units between adjacent resonances for the liquid and reference media, and f_q and Z_q respectively, are the frequency of the quartz and its acoustic impedance. Equation 13 may be approximated by a simpler expression (with a difference of only a few parts per thousand) as

$$\frac{\delta c_f}{c_f} = \frac{\delta f_n}{f_n} \qquad (14)$$

where δc_f is the velocity difference between the unknown and reference media, δf_n is the difference between corresponding resonances, and c_f is the velocity of the reference medium at a frequency f_n. In general, velocity measurements are more difficult to perform than absorption measurements due to temperature drift and other instabilities of the unit. The attenuation of the specimen solution may be calculated from the examination of the quality factor Q of the liquid filled cavity, defined as the frequency f_n divided by the 3 dB bandwidth, Δf_n, of the resonance. This quality factor is a function of the mechanical clamping and the attenuation per wavelength $\alpha\lambda$ of the ultrasonic energy and is expressed as,

$$Q = \frac{f_n}{\Delta f_n} = \frac{\pi}{\alpha\lambda} \qquad (15)$$

The measured Q, however, includes losses associated with attenuation in the solvent, as well as those arising from diffraction, wall effects, imperfect reflections at the quartz surface, etc.,

in addition to the desired excess attenuation due to the solute. Assuming all of these energy losses are additive [32], the measured quality factor Q is given by

$$\frac{1}{Q_{meas}} = \frac{1}{Q_{solute}} + \frac{1}{Q_{extra}} \qquad (16)$$

where Q_{solute} is the quality factor due to sound absorption in the solute, and Q_{extra} includes solvent and other cell losses previously discussed. The excess solute absorption can be obtained by means of a reference measurement in the same cell at the same frequencies with the reference liquid, having equal or very similar sound velocity to insure the same sound field pattern for both measurements. The excess absorption per wavelength in the specimen solution is then obtained from

$$(\alpha\lambda)_{excess} = \pi \left(\frac{\Delta f_s - \Delta f_r}{f_n} \right) \qquad (17)$$

where Δf_s and Δf_r are the corresponding 3 dB bandwidths of the n^{th} resonant peak in the sample and reference liquid, respectively. The application of an overpressure (usually about 10 psi) causes a slight concavity of the transducers, which reduces diffraction and boundary effects, and thereby reduces the minimum frequency of measurement [33]. This technique has been used in our laboratory over the frequency range from 0.5 to 10 MHz.

5. Absorption and Velocity Measurements of Shear Waves in Biological Specimens

A system based on a pulse superposition technique allows measurement of the shear acoustical properties of biological materials of interest [34]. The technique involves phase-amplitude balance to measure the magnitude and phase angle of the reflection coefficient of a shear wave impinging on a quartz-sample interface, and using the known impedance of the quartz, the complex shear specific impedance of the sample being investigated is obtained [35]. From the shear impedance and density, it is possible to calculate the dynamic shear stiffness μ_1, dynamic shear viscosity μ_2, shear velocity c_S, and shear absorption coefficient α_S as shown in the following relationships:

$$\mu_1 = \frac{R_S^2 - X_S^2}{\rho} \qquad (18)$$

$$\mu_2 = \frac{2R_S X_S}{\omega\rho} \qquad (19)$$

$$c_S = \frac{R_S^2 + X_S^2}{\rho R_S} \qquad (20)$$

$$\alpha_S = \frac{\rho\omega X_S}{R_S^2 + X_S^2} \qquad (21)$$

where ρ is the density, ω is the angular frequency, R_S is the real part of the specific shear acoustic impedance and X_S is the imaginary part of the specific shear acoustic impedance. This method has been used to measure the shear acoustic properties of tissues and other biological specimens [36-38]. An improved system utilizing a gated carrier as opposed to a pulsed oscillator approach and an improved ultrasonic unit of beveled AT quartz is being developed that will be useful over the frequency range from 2 to greater than 20 MHz and for temperatures from 10 to 40 °C. These and additional improvements should provide accuracies for the real and imaginary parts of the specific acoustic impedance within ± 100 mech-ohm/cm^2 at 10 MHz, allowing more accurate measurement of impedance close to that for water.

6. Concluding Remarks

The measuring techniques for ultrasonic absorption, attenuation and velocity in biological specimens discussed above have been only those utilized at the Bioacoustics Research Laboratory. There are, however, techniques employed elsewhere which may provide the same information in a more efficient manner. For example, the above discussed methods provide information at discrete frequencies and loci, or, for spatial averages. Spectrum analysis methods [39-41] have been promoted to permit measurements over a somewhat broadened frequency range. The measurement of the spatial distribution of ultrasonic properties such as attenuation or velocity within a specimen may be possible using the algebraic reconstruction from two-dimensional acoustic projections [42,43], although only velocity reconstruction has been considered practical so far.

Recent studies have also considered improvements regarding possible error in established methods of ultrasonic parameter measurements. Artifacts in acoustic attenuation due to phase cancellation effects have been described by Marcus and Carstensen [6], who compared a piezoelectric receiver (phase-sensitive) with a radiation-force receiver (phase-insensitive), and also by Busse et al. [44], comparing the piezoelectric receiver with an acousto-electric receiver. Both studies found that measurements using piezoelectric (phase-preserving) receivers yielded higher apparent attenuation values than those obtained using phase-insensitive receivers. Phase cancellation artifacts are thought to be the source of this error.

Acknowledgment

The authors acknowledge gratefully the partial support for portions of the activities described herein by grants from the National Institutes of Health.

References

[1] Rooney, J. A., Determination of acoustic power outputs in the microwatt-milliwatt range, Ultrasound Med. Biol. 1, 13-16 (1973).

[2] O'Brien, W. D., Jr., Ultrasonic Dosimetry, in Ultrasound: Its Application in Medicine and Biology, F. J. Fry, ed., Chap. 5 (Elsevier, Amsterdam, in press).

[3] Kossoff, G., Balance technique for the measurement of very low ultrasonic power outputs, J. Acoust. Soc. Am. 38, 880-881 (1965).

[4] Hill, C. R., Calibration of ultrasonic beams for biomedical applications, Phys. Med. Biol. 15, 241-248 (1970).

[5] Pohlman, R., Über die Absorption des Ultraschalls im menschlichen Gewebe und ihre Abhängigkeit von der Frequenz, Physik Z. 40, 159-161 (1939).

[6] Marcus, P. W. and Carstensen, E. L., Problems with absorption measurements of inhomogeneous solids, J. Acoust. Soc. Am. 58, 1334-1335 (1975).

[7] Dussik, K. T., Fritch, D. J., Kyriazidou, M., and Sear, R. S., Measurements of articular tissue with ultrasound, J. Phys. Med. 37, 160-165 (1958).

[8] Bauld, T. J. and Schwan, H. P., Attenuation and reflection of ultrasound in canine lung tissue, J. Acoust. Soc. Am. 56, 1630-1637 (1974).

[9] Wladimiroff, J. W., Craft, I. L., and Talbert, D. G., In vitro measurement of sound velocity in human fetal brain, Ultrasound Med. Biol. 1, 377-382 (1975).

[10] Buschmann, W., Voss, M., and Kemmerling, S., Acoustic properties of normal human orbit tissues, Ophtal. Res. 1, 354-364 (1970).

[11] Lange, J. N., Group velocity dispersion due to pulse reflection from a frequency dependent boundary impedance, J. Acoust. Soc. Am. 40, 998-1001 (1966).

[12] Goldman, D. E. and Richards, J. R., Measurement of high frequency sound velocity in mammalian soft tissues, J. Acoust. Soc. Am. 26, 981-983 (1954).

[13] Hueter, T. F., Viscoelastic Losses in Tissues in the Ultrasonic Range, Wright Air Development Center Technical Report 57-706 (1958).

[14] Fry, W. J. and Dunn, F., Ultrasound: Analysis and Experimental Methods in Biological Research, in Physical Techniques in Biological Research, W. L. Nastuk, ed., Vol. IV, Chap. 6, pp. 261-394 (Academic Press, New York, 1962).

[15] Dunn, F. and Fry, W. J., Ultrasonic absorption and reflection by lung tissue, Phys. Med. Biol. 5, 401-410 (1961).

[16] Fry, W. J. and Fry, R. B., Determination of absolute sound levels and acoustic absorption coefficients by thermocouple probes--theory, J. Acoust. Soc. Am. 26, 294-310 (1954).

[17] Fry, W. J. and Fry, R. B., Determination of absolute sound levels and acoustic absorption coefficients by thermocouple probes--experiment, J. Acoust. Soc. Am. 26, 311-317 (1954).

[18] Dunn, F., Temperature and amplitude dependence of acoustic absorption in tissue, J. Acoust. Soc. Am. 34, 1545-1547 (1962).

[19] Dunn, F. and Brady, J. K., Temperature and Frequency Dependence of Ultrasonic Absorption in Tissue, in Proceedings 8th International Congress on Acoustics, Vol. 1, p. 366c (Goldcrest Press, Trowbridge, Wilts., 1974).

[20] Brady, J. K., Goss, S. A., Johnston, R. L., O'Brien, W. D., Jr., and Dunn, F., Ultrasonic propagation properties of mammalian testes, J. Acoust. Soc. Am. 60, 1407-1409 (1976).

[21] Dunn, F. and Breyer, J. E., Generation and detection of ultra-high-frequency sound in liquids, J. Acoust. Soc. Am. 34, 775-778 (1962).

[22] Carstensen, E. L., Li, K., and Schwan, H. P., Determination of the acoustic properties of blood and its components, J. Acoust. Soc. Am. 25, 286-289 (1953).

[23] Pauly, H. and Schwan, H. P., Mechanism of absorption of ultrasound in liver tissue, J. Acoust. Soc. Am. 50, 692-699 (1971).

[24] O'Brien, W. D., Jr., The Absorption of Ultrasound in Aqueous Solutions of Biological Polymers, Ph.D. Thesis, University of Illinois, Urbana, Illinois (1970).

[25] Goss, S. A., The Concentration Dependence of the Ultrasonic Absorption in Aqueous Solutions of Bovine Serum Albumin, M.S. Thesis, University of Illinois, Urbana, Illinois (1974).

[26] Kessler, L. W., Hawley, S. A., and Dunn, F., Semi-automatic determination of ultrasonic velocity and absorption in liquids, Acustica 24, 105-107 (1971).

[27] Pellam, J. R. and Galt, J. K., Ultrasonic propagation in liquids: I. Application of pulse technique to velocity and absorption measurements at 15 megacycles, J. Chem. Phys. 14, 608-614 (1946).

[28] Del Grosso, V.A., Systematic Errors in Ultrasonic Propagation Parameter Measurements Part I--Effect of Free-Field Diffraction, U.S. Naval Research Laboratory Report 6026 (1964).

[29] Schwan, H. D. and Carstensen, E. L., Ultrasonics aids diathermy experiments, Electronics 25, 216-220 (1952).

[30] Carstensen, E. L., Measurement of dispersion of velocity of sound in liquids, J. Acoust. Soc. Am. 26, 858-861 (1954).

[31] Eggers, F., Eine Resonatormethode zur Be-

stimmung von Schall--Geschwindigkeit und Dämpfung an geringen Flüssigkeitsmengen, Acustica 19, 323-329 (1968).

[32] Eggers, F. and Funck, Th., Ultrasonic measurements with milliliter liquid samples in the 0.5-100 MHz range, Rev. Sci. Instrum. 44, 969-977 (1973).

[33] Eggers, F., Funck, Th., and Richman, K. H., High Q ultrasonic liquid resonators with concave transducers, Rev. Sci. Instrum. 47, 361-367 (1976).

[34] Frizzell, L. A., Carstensen, E. L., and Dyro, J. F., Shear properties of mammalian tissues at low megahertz frequencies, J. Acoust. Soc. Am. 60, 1409-1411 (1976).

[35] Moore, R. S. and McSkimin, H. J., Dynamic Shear Properties of Solvents and Polystyrene Solutions from 20 to 300 MHz, in Physical Acoustics, W. P. Mason and R. N. Thurston, eds., Vol. VI, Chap. 4, pp. 167-242 (Academic Press, Inc., New York, 1970).

[36] Dyro, J. F., Ultrasonic Study of Material Related to Atherosclerotic Plaque--Dynamic Viscoelastic Properties of Cholestoric Esters, Ph.D. Thesis, University of Pennsylvania, Philadelphia, Pennsylvania (1972).

[37] Dyro, J. F. and Edmonds, P. D., Ultrasonic absorption and dispersion in cholesteryl esters, Mol. Cryst. Liq. Cryst. 25, 175-193 (1974).

[38] Frizzell, L. A., Ultrasonic Heating of Tissues, Ph.D. Thesis, University of Rochester, New York (1975).

[39] Chivers, R. C. and Hill, C. R., Ultrasonic attenuation in human tissue, Ultrasound Med. Biol. 2, 25-29 (1975).

[40] Lizzi, F., Katz, L., St. Louis, L., and Coleman, D. J., Applications of Spectral Analysis in Medical Ultrasonography, Ultrasonics 14, 77-80 (1976).

[41] Le Croissette, D. H. and Heyser, R. C., Attenuation and Velocity Measurements in Tissue Using Time Delay Spectrometry, in Ultrasonic Tissue Characterization, M. Linzer, ed., National Bureau of Standards Spec. Publ. 453, pp. 81-95 (U.S. Government Printing Office, Washington, D.C., 1976).

[42] Greenleaf, J. F., Johnson, S. A., Lee, S. L., Herman, G. T., and Wood, E. H., Algebraic Reconstruction of Spatial Distributions of Acoustic Absorption Within Tissue From Their Two-Dimensional Acoustic Projections, in Acoustical Holography, P. S. Green, ed., Vol. 5, pp. 591-603 (Plenum Press, New York, 1974).

[43] Greenleaf, J. F., Johnson, S. A., Samayoa, W. F., and Duck, F. A., Algebraic Reconstruction of Spatial Distributions of Acoustic Velocities in Tissue From Their Time of Flight Profiles, in Acoustical Holography, N. Booth, ed., Vol. 6, pp. 71-90 (Plenum Press, New York, 1975).

[44] Busse, L. J., Miller, J. G., Yuhas, D. E., Mimbs, J. W., Weiss, A. N., and Sobel, B. E., Phase Cancellation Effects: A Source of Attenuation Artifact Eliminated by a CdS Acoustoelectric Receiver, in Ultrasound in Medicine, D. White, Ed., Vol. 3, pp. 1519-1535 (Plenum Press, New York, 1977).