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Shear properties of mammalian tissues at low megahertz frequencies*

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Rough values for the transverse-wave specific acoustic impedance $Z_s = R_s + jX_s$, transverse velocity of sound c_s, and transverse absorption coefficient α_s have been measured for canine liver, kidney, muscle tissues, and for packed red blood cells in the frequency range from 2 to 14 MHz at 25°C. The ranges of the results are $R_s = 700$–$3000\ \Omega_{mech}/cm^2$, $X_s = 400$–$4000\ \Omega_{mech}/cm^2$, $c_s = 900$–10000 cm/sec, and $\alpha_s = 2000$–30000 np/cm. The corresponding results for shear stiffness μ_1 and viscosity μ_2 are $\mu_1 < 10^7$ dyn/cm^2 and $\mu_2 = 4$–30 centipoise. At these frequencies the viscosities are orders of magnitude less than those reported at 0.5–5 kHz by Oestreicher (1951). The low impedances (viscosities) correspond to low velocities and extremely high absorption coefficients for shear waves in tissue.

Subject Classification: [43]80.20, [43]80.30; [43]35.26; [43]85.20.

INTRODUCTION

It has been recognized for many years that transverse waves may play an important part in the interaction of ultrasound with tissue (Carstensen, 1960). For lack of data, however, only speculations have been possible (Schwan and Carstensen, 1952; Lehmann, 1953; Baldes *et al.*, 1958; Taylor and Connolly, 1969; O'Brien *et al.*, 1972). For some insight into these phenomena, rough measurements of the shear acoustic impedance of several mammalian tissues have been made at low megahertz frequencies. Transverse acoustic properties of tissue at these frequencies appear to be determined primarily by the stiffnesses and viscosities of the subcellular constituents.

I. EXPERIMENTAL METHOD

A. Equipment

A pulse superposition technique as developed by McSkimin (1961) was used to measure the complex shear specific acoustic impedance. The technique involved measurement of the magnitude and phase angle of the reflection coefficient of a shear wave impinging on a quartz-sample interface. This was compared with total reflection from a quartz–air interface. From a knowledge of the impedance of the quartz, one then obtained the complex impedance of the sample being investigated (McSkimin, 1961). From the shear impedance and density, it was 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:

$$\mu_1 = (R_s^2 - X_s^2)/\rho\,, \tag{1}$$

$$\mu_2 = 2R_sX_s/\omega\rho\,, \tag{2}$$

$$c_s = (R_s^2 + X_s^2)/\rho R_s\,, \tag{3}$$

$$\alpha_s = \rho\omega X_s/(R_s^2 + X_s^2)\,, \tag{4}$$

where ρ is the density, ω is the angular frequency, R_s is the real part of the specific shear acoustic impedance

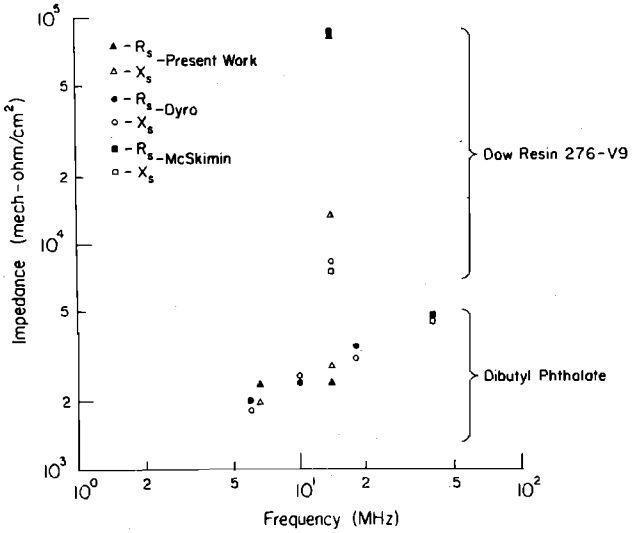

FIG. 1. Test of measuring system. Measurements of dibutyl phthalate and Dow Resin 276-V9 were made and compared to earlier measurements by Dyro (1972), McSkimin (1956), and McSkimin and Andreatch (1967).

and X_s is the imaginary part of the specific shear acoustic impedance. A more complete description of the technique and equipment used is given by Dyro (1972), Dyro and Edmonds (1975), and Frizzell (1975).

Sample temperature was measured with a copper–Constantan thermocouple and Bailey BAT-4 Thermocouple Thermometer. Temperature stabilization during an experiment was maintained within ±0.05 °C. Temperature from experiment to experiment ranged from 24.8 to 26.3 °C.

B. Coupling

Because of the extremely small depths of penetration of shear waves, one of the greatest concerns with solid samples was whether the measurements were a true indication of the properties of a solid, or whether they were dominated by a coupling layer. The best coupling appeared to be obtained by pressing the sample, under moderate pressure, into direct contact with the buffer rod. In some liver and kidney samples, the capsular surface was wiped dry and pressed against the buffer rod. The results of these measurements were essentially the same as those obtained when a cut surface of the tissues was pressed against the rod.[1]

As a test of this coupling procedure, measurements were made with a potting compound (General Electric Co. RTV 615). The sample was first measured in the liquid state, then allowed to harden on the rod. In the process a weak bond is formed between the quartz and the RTV. The sample was removed, then pressed to the rod in the manner used for tissue samples. The results are shown in Table I. Although the resistance of the RTV clearly changes on hardening, there was essentially no difference between measurements of samples which had solidified on the rod and those pressed to the rod.

C. Comparisons with previous measurements

Measurements were made on dibutyl phthalate and Dow Resin 276-V9 and compared to measurements reported earlier by McSkimin (1956), McSkimin and Andreatch (1967), and Dyro (1972). Good agreement between our observations and earlier data is shown in Fig. 1.

D. Errors

The tissue impedances were so low that the measuring system was taxed to the limit of its sensitivity in these studies. The uncertainties in the value of R_s were in the order of $\pm 10^3\ \Omega_{mech}/cm^2$. Errors in determination of X_s were generally in the order of $\pm 2.5\times10^3\ \Omega_{mech}/cm^2$ for tissues and water.

For samples reported in this study, calculations of stiffness involved a difference between numbers of the same order of magnitude [Eq. (1)]. As a result of this and system limitations, it was impossible to measure stiffness below about 10^7 dyn/cm^2. The stiffness of all samples measured was below this value. To put this in perspective, the values of shear stiffness of tissues are much smaller than the values of 2.6×10^9 dyn/cm^2 for polyethylene and 2.5×10^{11} dyn/cm^2 for aluminum.

II. SAMPLE PREPARATION

Fresh tissues were excised from dog, placed in cold saline, and either sliced for measurement within minutes of excision or stored at 4 °C. Storage time ranged from several hours to several days. Immediately before measurement whole tissue samples were sliced 3–4 mm thick and about 2.5 cm in diameter and placed in the sample holder.

Liver homogenate was prepared by cutting whole liver and then chopping the liver in a blender until it was reduced to a subcellular homogenate as determined by microscopic examination. The samples were then centrifuged at 4000 rpm in a Sorvall centrifuge for 4 min to remove air bubbles introduced during homogenation.

TABLE I. Test of coupling technqiue. General Electric Co. RTV 615 was placed in sample holder in liquid form and allowed to harden on quartz buffer rod. Measurements in this condition are compared to those after the sample was removed and pressed in contact with the rod. The standard deviations are indicated.

Frequency	Sample	Real part of impedance $\left(\frac{\Omega_{mech}}{cm^2}\right)$	Imaginary part of impedance $\left(\frac{\Omega_{mech}}{cm^2}\right)$
14	Liquid	5500 ± 1500	3700 ± 800
14	Solid-hardened on quartz	9600 ± 1500	4600 ± 2000
14	Solid-press contact	8300 ± 1100	3700 ± 1400

TABLE II. Summary of the shear wave measurements from this investigation. The range of values includes measurements on water, Soni Gel, red blood cells, liver, liver homogenate, kidney, and muscle at 2.0, 6.5, and 14 MHz. A density of 1 g/cm^3 was used for all samples.

Parameter	Range of Values	Units
Real part of impedance	$(0.7-3)\times10^3$	Ω_{mech}/cm^2
Imaginary part of impedance	$(0.4-4)\times10^3$	Ω_{mech}/cm^2
Stiffness	$<1\times10^7$	dyn/cm^2
Viscosity	4–30	cP
Velocity	$(0.9-10)\times10^3$	cm/sec
Absorption coefficient	$(2-30)\times10^3$	cm^{-1}

III. RESULTS

Order of magnitude data on the properties of shear waves in liver, homogenized liver, kidney, muscle, red blood cells, Soni-Gel, and water have been obtained in the 2–14-MHz frequency range. Table II reports the ranges of values for each of the measured and computed properties. The ranges were derived by combining the data at all frequencies on all samples. Although the data provide only upper limits to the stiffness and order of magnitude values for viscosity, shear velocity, and shear absorption coefficient, the values in Table II are the first measurements of the shear properties of soft tissues in the low megahertz frequency range, and as such will permit quantitative calculations of the importance of mode conversion at tissue interfaces. There are several points in these data worth noting:

(1) The shear properties of tissues, packed red cells, and noncellular tissue homogenates were of the same order of magnitude.

(2) At low megahertz frequencies the values of viscosity are much lower than those reported at very low and zero frequency. The value of the order to 20 centipoise for packed red blood cells (hematocrit = 85) is low when compared to a value greater than 100 centipoise at low shear rates and zero frequency (Brooks, Goodwin and Seaman, 1970).

(3) Wavelengths calculated from these measurements (3 – 7.5 μm) are generally on the order of the cell size (Albritton, 1952; Windle, 1960).

(4) The low impedances (low viscosities) measured for these samples correspond to extremely high absorption coefficients for tissues.

IV. DISCUSSION

Oestreicher (1951) showed that in the frequency range from 500 to 5000 Hz the shear properties of tissue were essentially those of a fluid with a density of 1.1 g/cm^3, viscosity of 150 poise and zero stiffness. The present study at megahertz frequencies shows that the magnitudes of the viscosities (4 to 30 centipoise) are orders of magnitude lower than those measured by Oestreicher at low frequencies. It is suggested that this dramatic decrease in the dynamic viscosity of tissues is due to a shift in the level of tissue components controlling the shear properties. At low frequencies the shear acoustic properties are determined to a high degree by the cellular structure, whereas at high frequencies with shorter wavelengths the macromolecular composition of the tissue becomes more important. It has been recognized for years that the propagation of longitudinal waves in tissues is controlled primarily at the macromolecular rather than the cellular level (Carstensen, 1960). The results of this study suggest that the same is probably true for transverse waves.

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‡In attempts to use a tissue cement to bond samples to the rod, it appeared that the impedance measured was just that of the adhesive.

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