Temperature and Amplitude Dependence of Acoustic Absorption in Tissue

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The acoustic intensity absorption coefficient of tissue of the central nervous system has been determined at the sound frequency of 1 Mc/sec in the temperature range from 2° to 28°C at incident sound intensities ranging from 5 to 200 W/cm². The absorption coefficient exhibits an increase with increasing temperature and no variation with the acoustic intensity.

IT is well established that properly controlled, high-intensity ultrasound produces unique effects on the central nervous system[1] and provides a versatile method for modifying brain structures.[2] In order that the full potentialities of this methodology be realized for fundamental biological research and for medicine, it is essential that the physical mechanism(s) of the interaction of intense ultrasound and biological structures be understood. An important adjunct to the elucidation of these physical mechanisms is a basic understanding of the absorption processes occurring when biological materials are irradiated with ultrasound. The frequency dependence of the absorption coefficient of most investigated tissues can be described by a power function whose exponent varies between 1 and 1.3.[3,4] Knowledge of the temperature dependence of the ultrasonic absorption coefficient of tissue is lacking, owing to the fact that investigations have been restricted largely to adult mammals, which are thermally homeostatic, and thereby essentially excluding the possibility of making absorption measurements over a wide temperature range. The present study was undertaken to fill, in part, this existing gap in knowledge and to provide necessary data for the accurate determination of the dosage parameters to affect specific biological systems.

The method employed for determining the value of absorption coefficients in tissue has been described previously.[3,5,6] Briefly, a small calibrated thermocouple probe is imbedded in the tissue and the specimen is exposed to rectangular acoustic pulses of known intensities. The transient thermoelectric output produced in response to a pulse is recorded on a magnetic oscillograph employing a galvanometer with a time constant of approximately 0.02 sec. The rise time of the pulse is of the order of 10^{-3} sec. The transient temperature change detected by the thermocouple imbedded in the tissue possesses two distinct phases. The first phase, which reaches an equilibrium value rapidly (in about 0.1 sec), results from the conversion of acoustic energy into heat by the viscous forces acting between the wire and the immediately surrounding medium.[7] This phase is, of course, not present in the tissue when the thermocouple is absent. The second phase exhibits an almost linear characteristic (for a pulse duration of approximately 1 sec) and results from acoustic energy converted into heat by absorption in the tissue surrounding the thermocouple junction. If the thermocouple wires are sufficiently small in diameter, the initial time rate of change of temperature from the second phase is related to the acoustic intensity absorption coefficient per unit path length by the relation

$$\mu = (\rho C_p K/I)(dT/dt)_0, \tag{1}$$

where ρC_p is the heat capacity per unit volume of the tissue (cal cm^{-3} C), I is the acoustic intensity (W/cm²), and K is the mechanical equivalent of heat. The acoustic intensity of the plane traveling wave field is determined by a thermoelectric probe which has been calibrated against a radiation-pressure detector.[3,7,8]

The young mouse, 24 h after birth, is a convenient preparation for the study described herein for a number of reasons,[5,9,10] one of the more important being that it is an essentially poikilothermic animal which readily allows temperature cycles to as low as 0°C to be carried out without producing permanent changes in the animal. The following procedure is followed in preparing the animals for ultrasonic irradiation and the associated measurements of the concomitant transient temperature rise.[5] The animal is cooled to render it dormant so that it can be properly positioned in the mouse holder and to ensure that it will remain in that position until it is placed in the sound tank and irradiated. The traveling-wave sound field is produced by a 1-Mc, unfocused, X-cut quartz plate. The acoustic transmission medium is degassed mammalian Ringer's solution. When the mouse is sufficiently cool, it is placed in the mouse holder which secures the head, hind limbs, and tail firmly. The previously fabricated and calibrated copper-constantan thermocouple is then inserted in the spinal

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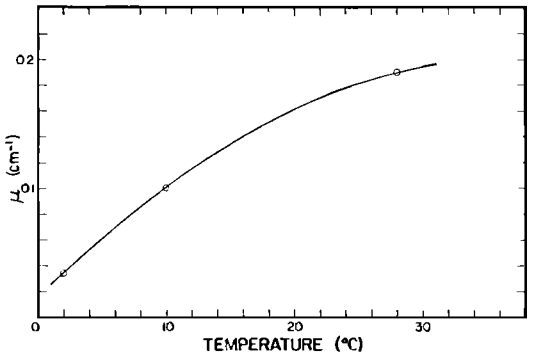

FIG. 1. Acoustic intensity absorption coefficient vs temperature at 1 Mc for spinal cord of young mice.

cord at the level of the third lumbar vertebra. (There is very nearly a 1:1 spatial correspondence between vertebral segments and cord segments of these animals.) An optical arrangement is employed to position the animal accurately, and hence the junction of the thermocouple, in the predetermined site in the sound field. The method of inserting the thermocouple into the tissue (threading) does not yield information about the precise position of the junction with respect to the anatomical structure of the animal. This is determined after the animal is sacrificed. The soft tissues are cleared in a 1% aqueous solution of KOH after which the oseous material is stained with Alizarin Red-S.[11] The specimen can then be viewed under a microscope and the position of the thermocouple junction is located accurately with respect to the vertebral structures.

The thermocouples are fabricated from 0.003-in.-diameter commercially available wires, which are etched in acid to reduce the diameter to approximately 0.0005 in. in the vicinity of the junction. Assembly of the thermocouple is accomplished by soldering. Both lapp and butt joints have been used without any observed difference in the ensuing results.

Table I shows the results at the three base temperatures of the animals considered in this study. The base temperatures of the animals were known to ±0.1°C. The quantity $\mu/\rho C_p$ is computed using Eq. (1) and a knowledge of the incident sound intensity and the experimentally determined initial temperature rise measured by the inserted thermocouple (the temperature rise associated with the viscous forces being subtracted away). The acoustic intensity absorption coefficient per unit path length, μ, is computed using the value[12] $\rho C_p = 0.84$ cal/cm³ °C for the heat capacity per unit volume ($\rho = 1.03$ g/cm³, $C_p = 0.81$ cal/g°C). In the absence of more specific information, the heat capacity per unit volume is considered to be constant within the temperature range of these experiments. The over-all uncertainty in the numerical values of $\mu/\rho C_p$ is thought to be of the order of 10–15% so that a small dependence upon temperature of ρC_p does not add appreciably to the percentage uncertainty in the results. It should be noted that in computing the values of $\mu/\rho C_p$ and μ, account was taken of the reduction in the incident intensity caused by absorption of energy in the tissue. This small correction was accomplished by using a value of μ obtained by inserting the value of the incident intensity in Eq. (1) and this is considered to be sufficiently accurate for the purposes of this study. Figure 1 is a graphical representation of the intensity absorption coefficient versus temperature.

TABLE I. Absorption data.

T(°C)	$\mu/\rho C_p$	μ(cm^{-1})
2	0.040	0.034
10	0.12	0.10
28	0.23	0.19

The procedure for acquiring the data included the exposure of each specimen to different levels of incident sound. Thus data is available relating the initial time rate of change of temperature, as observed by the inserted thermocouple probe, as a function of the intensity of the incident acoustic wave and is illustrated graphically in Fig. 2. On the assumption that the heat capacity per unit volume, ρC_p, is not dependent upon the intensity, it can be concluded [see Eq. (1)] that, in the range of incident sound intensities from zero to approximately 200 W/cm², the absorption coefficient is independent of the intensity of the incident wave for the experimental preparations of this study. Concerning the observed lack of dependence of the absorption coefficient of nerve tissue upon the sound intensity in these experiments, the following statements can be made: The propagation distance from sound source via degassed mammalian saline to tissue was, in these experiments, approximately 5 cm. The discontinuity distance in water for the highest sound amplitudes employed in these experiments is approximately 8 to 10 cm. The

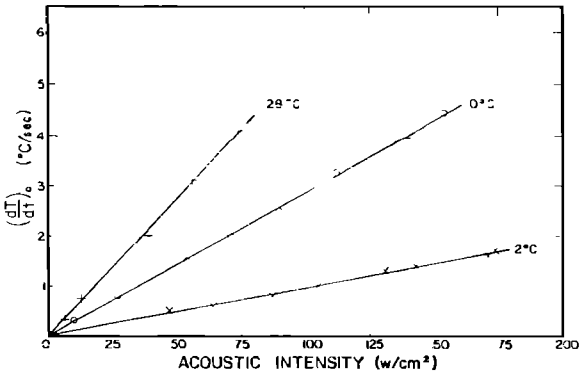

FIG. 2. Time rate of change of temperature in spinal cord of young mice produced by absorption of 1 Mc sound vs incident acoustic intensity.

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product αL (the infinitesimal amplitude absorption coefficient × the discontinuity distance) is of the order of 10^{-3}.[3] Using the results of recent computations,[13] it is seen that the amplitude of the fundamental frequency component of the propagated sound wave at the tissue - saline interface is at most a few percent less than that initially at the surface of the vibrating element. Thus the wave incident at the tissue is relatively undistorted. On the assumption that the ratio B/A for tissue is approximately the same as that for water, and since the propagation distance in the tissue is relatively short (approximately 0.5 mm), the transfer of energy from the incident fundamental wave to its harmonics is negligible.

The form of the temperature dependence of the absorption coefficient (Fig. 1) eliminates shear viscosity as the absorption mechanism in tissue. On the basis of the work reported herein the following empirical relation describes the temperature dependence of the amplitude absorption coefficient in the range from 0° to 35°C:

$$\alpha=\tfrac{1}{10}[2-e^{0.016(35-T)}], \qquad (2)$$

where T is the temperature of the tissue in °C. The temperature dependence of the absorption coefficient appears to resemble that of high viscosity liquids above the main relaxation frequencies of the liquid.[14] Indeed, it has been estimated that the bulk viscosity of muscle relaxes near 40 kc and the shear viscosity near 400 kc.[15]

ACKNOWLEDGMENT

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Optical Effects of Ultrasonic Waves Producing Phase and Amplitude Modulation

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A theory is developed for the diffraction of light by ultrasonic waves of sufficiently high frequency, large amplitude, and/or large beamwidth that the emerging light wavefront is significantly amplitude modulated in addition to the phase modulation considered in the Raman–Nath approach. The ultrasonic beam is considered to consist of N adjacent sections and the final diffraction spectrum to result from N successive diffractions. The diffraction orders emerging from a given section are considered to be sources for further diffraction by the next section. Only phase modulation of the separate plane waves (diffraction orders) is considered within a section. Refraction of light is not considered as such; it is characterized by successive redistribution of light in the diffraction orders. Numerical results are obtained by iterations using an electronic computer. These results are compared with measurements.

INTRODUCTION

THE diffraction of light by ultrasonic waves has been the subject of numerous experimental and theoretical investigations. Raman and Nath[1] developed a theory for conditions under which the ultrasonic frequency, amplitude, and/or beamwidth are sufficiently small that the *refraction* of light within the ultrasonic beam may be neglected. Under such conditions the ultrasonic beam may be considered to act as a pure *phase* grating producing only changes in the relative phase of the initially plane incident light wave and no intensity changes. The Raman–Nath approach has proven useful over a limited range. Herein, the Raman–Nath theory and other theories based on the phase grating concept are called the Raman–Nath approach for sake of brevity.

When the ultrasonic frequency, amplitude, and, or beamwidth become large, refraction causes significant amplitude modulation along the emerging light wavefront. Extermann and Wannier,[2] Wagner,[3] Van Cittert,[4] and Mertens[5] have obtained solutions for such conditions. Their results contain varying degrees of approximation and complexity.

In this paper, a solution to the problem of diffraction of light by sinusoidal, plane, progressive, ultrasonic waves is presented. The ultrasonic beam is considered to consist of N adjacent ultrasonic beams. For N

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