# ACOUSTIC-REFLEX EFFECTS ON THE INPUT ADMITTANCE AND TRANSFER CHARACTERISTICS OF THE HUMAN MIDDLE-EAR

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## ABSTRACT

In the past 50 years many kinds of experiments on animals have demonstrated that muscles in the middle-ear contract in response to intense sound and change the input admittance and transmission of the middle-ear. Measurements of acoustic-reflex effects on input admittance in humans are now being used clinically for various diagnostic purposes. Most measurements of transmission-change in humans have been rather indirect. The goal of this thesis was to measure transmission-change (using a new psychophysical technique) and input admittance-change in normal subjects to determine what these changes are and what relationships they have to each other.

An acoustic device was constructed for obtaining the acoustic admittance (or impedance) at the eardrum of normal human ears; the device was demonstrated to provide valid measurements of absolute admittance (magnitude and angle) from 62 Hz to 4 kHz. A computer-controlled system was used to control stimuli and to collect and manipulate data. Measurements were made for 61 discrete frequencies (10 frequencies per octave) in about 73 sec. Special attention is given to estimating the earcanal space $V_C$ between the eardrum and the tip of the measuring device. To determine $V_C$, admittance measurements with earcanal static pressures of +40 and -40 cm $H_2O$ (re ambient) are used; however, in contrast to the usual assumption applied in clinical tympanometry, we do not assume these static pressures reduce the admittance at the eardrum to zero. Results are reported for four subjects. Below 500 Hz, the eardrum admittance is compliance dominated; our average equivalent compliance of 0.70 cc is in agreement with values reported by others. From 1 to 4 kHz, the resistance component of the eardrum impedance exceeds the magnitude of the reactance component; our resistance values exceed those reported by others, but are in general agreement with values suggested from analysis of reported acoustical measurements on the external ear.

Psychophysical measurements of middle-ear transmission-change and physical measurements of input admittance-change are reported for the same four subjects during activation of acoustic reflexes with contralateral band-pass noise at levels up to 110 dB SPL. The psychophysical data have been obtained (from 0.4 to 4 kHz) using the sensitivity of the phase of the $2f_1-f_2$ aural combination tone to stimulus level. As a function of the reflex-elicitor level, $L_N$, admittance- and transmission-change exhibit approximately equal "reflex thresholds" of $L_N$ = 85 to 90 dB SPL; however, for $L_N > 105$ dB SPL, admittance-change nearly saturates whereas transmission-change continues to increase. For $L_N$ = 110 dB SPL, the average low-frequency admittance-change is -2.5 dB ($\sigma$ = 1.4 dB); the average low-frequency transmission-change is -10 dB ($\sigma$ = 1.8 dB). The results are interpreted in terms of changes in the linear-incremental properties of the mechanics of the middle-ear system.

THESIS SUPERVISOR: William T. Peake
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## Organizational note

This thesis has been written in two parts. Each part is in a format (albeit not a length) intended to be compatible with papers submitted to the Journal of the Acoustical Society of America. This has been done as part of the continuing effort by the Auditory Perception Group at M.I.T. to decrease the effort and delay traditionally associated with getting thesis research submitted for publication.

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## PART 1.

## PART 2.

# PART 1.

# ON THE INPUT ACOUSTIC ADMITTANCE OF THE HUMAN MIDDLE-EAR

## I. Introduction

Knowledge of the acoustic admittance (or impedance) at the eardrum is of interest because it may (1) provide insight into basic mechanisms of the middle ear, (2) aid in diagnosis of some pathologies of the ear, and (3) lead to standards for earphone couplers (c.f., Møller, 1960, 1961; Zwislocki, 1962, 1970, 1971; Sachs and Burkhard, 1972b; Lilly, 1973; Shaw, 1974b, 1975a, 1975b; Jerger, 1975; Feldman and Wilber, 1976; Northern, 1976). Although numerous studies of the acoustic admittance at the eardrum in humans have been published, "reliable" measurements have not been reported for frequencies above 1.5 kHz (Shaw, 1975b, pg. 38). The factor which appears to be primarily responsible for this limitation is that the geometry of the earcanal and sensitivity to pain associated with mechanical stimulation of its inner end make it difficult to perform measurements "at the eardrum"; hence, the admittance measurements are typically obtained at some distance from the eardrum. Accurate specification of the space between the locus of measurement and the eardrum is difficult and is of crucial importance in transforming the measured admittance to estimates of the admittance at the eardrum.

Herein we present a method for obtaining the admittance at the eardrum in normal human ears and demonstrate its validity for frequencies up to 4 kHz. The acoustic device used is similar in principle to that used in some commercially-available, "clinical" admittance-measuring devices. By calibrating the device with known acoustic admittances, we can obtain measurements of the absolute admittance (magnitude and angle) over a wide range of frequencies. A computer-controlled system is used to obtain and store the measurements; this permits the measurements to be performed at many frequencies rapidly and also facilitates manipulation of the data.

The estimation of the "residual earcanal space" between the eardrum and the tip of the measurement probe in the earcanal is given special attention and in part makes use of admittance measurements obtained with static pressures applied in the earcanal. However, in contrast to the usual assumption applied in clinical measurements, we do not assume that the applied static pressure reduces the admittance at the eardrum to zero.

The method presented herein has also been applied to study the effects of the acoustic reflex and of static pressure on the admittance at the eardrum. Such admittance measurements along with measurements of the effects on middle-ear transmission are reported for the acoustic reflex in Part 2 of this thesis. Results from similar measurements of the effects of static pressure are in preparation.

## II. Methods and preliminary results

### II. A. Scheme for admittance measurements

Admittance measurements at the tip of a probe inserted into the earcanal were accomplished using a specially designed electroacoustic-probe-assembly (EPA) system.[1] The EPA (see Fig. 1) incorporates three subsystems: (a) a source of volume velocity (an earphone and coupling tube), (b) a sound-pressure indicator [a probe-tube/microphone (MIC) combination], and (c) a static (DC) pressure system. These three systems converge at an earpiece unit with three channels. An inflatable cuff fits over the earpiece and is used to hermetically seal the EPA in the earcanal.

The channels for the volume velocity and the static pressure end flush with the earpiece tip; the microphone probe-tube, however, extends beyond the earpiece tip. This extension is required to obtain "valid" measurements above ~2.5 kHz. Specifically, we desire to measure the sound pressure resulting only from the longitudinal uniform plane-wave in the earcanal. At frequencies above ~2.5 kHz, however, other modes contribute significantly to the sound pressure in the immediate vicinity of the volume-velocity inlet. The probe-tube "extension" enables one to sample the sound pressure beyond the region "contaminated" with these modes. [Sec. II. E includes measurements aimed at empirically evaluating this problem; further treatment of the acoustics is given by Ingard (1948) and Sachs and Burkhard (1972a).] The extension which could be used varied from subject to subject (from 0 to 4 mm, see Sec. II. H).

The terminal characteristics of the EPA can be represented conveniently by the Norton equivalent model (c.f., Bose and Stevens, 1965, pp. 301-306) shown in the lower right of Fig. 1. A sinusoidal input voltage to the EPA of complex amplitude[2] $V_i$ leads to a Norton volume velocity source of $U_t = K_i V_i$, where $K_i$ represents the overall transformation of the earphone driver, the earphone, and the coupling tube, and the earpiece unit. A portion of $U_t$ designated $U_\ell$ is delivered to the load admittance $Y_\ell$ seen "looking out" from the plane of the EPA-tip. The remaining volume velocity, $U_t - U_\ell$, is delivered to the "self" (i.e. "internal") admittance of the EPA represented by the source admittance $Y_s$. The two admittances $Y_\ell$ and $Y_s$ are "in parallel" and sum to $Y_t$. This admittance and the Norton volume velocity determine the sound pressure $P_t$ which is developed at the EPA-tip, $P_t = U_t/Y_t$. $P_t$ is measured through the probe-tube/microphone combination, with transfer function $K_{pm}$, as the voltage $V_{pm} = K_{pm} P_t$. Following amplification, $V_{pm}$ appears as the EPA output voltage $V_o$. [We have modeled the acoustics as if $P_t$, $U_t$, and $Y_t$ are all in the plane of the EPA-tip, neglecting the probe-tube extension. A more detailed model with specific accounting for the sound-pressure and volume-velocity differences (for the longitudinal mode) between the EPA-tip and the extended probe-tube tip did not seem waranted, since for the maximum extension used (4 mm), these two positions are separated by less than one twentieth of a wavelength of sound at 4 kHz.]

To calibrate the EPA for admittance measurements, two quantities must be determined: the voltage ratio $[V_{pm}/V_i]_{REF}$ for a known "reference" admittance $Y_{\ell\text{-REF}}$, and the quantity $Y_s$. An unknown load admittance $Y_\ell$ can then be determined from measured values of $V_i$ and $V_{pm}$ as:

$$Y_\ell = \frac{V_i}{V_{pm}} \left\{ \left[ \frac{V_{pm}}{V_i} \right]_{REF} \bullet \left[ Y_s + Y_{\ell\text{-REF}} \right] \right\} - Y_s \qquad (1)$$

To determine $Y_s$, measurements of $V_i$ and $V_{pm}$ for two known loads are used and Eq. 1 is solved for $Y_s$.

Notice that the transfer functions $K_i$ and $K_{pm}$ do not appear in Eq. 1; they are not explicitly needed for admittance measurement. However, knowledge of these transfer functions is useful for the control of (using $K_i$) and calculation of (using $K_{pm}$) the sound pressure at the EPA-tip. These transfer functions were determined with the aid of a supplementary measurement with a reference microphone used to measure $P_t$ with a known $V_i$, $V_{pm}$, and $Y_\ell$ (see Sec. II. E).

For use as known acoustic admittances, three precision cavities were fabricated. Each cavity was cylindrical and the EPA earpiece, with the inflatable cuff removed, press fit along the axis of each cavity so that the earpiece tip was flush with the inside surface of the cavity. The three cavities were machined from plexiglass and were of nominal volumes 0.5, 1.0, and 2.0 cc (and exact volumes 0.51, 1.01, and 1.96 cc and lengths, respectively, 0.51, 1.01, and 1.01 cm). These particular test loads were chosen because their admittance magnitudes are in the approximate range of admittance magnitude expected to exist in the earcanal, and additionally, because their admittances can be theoretically specified. Ignoring viscous and thermal losses, assuming the cavity walls are rigid, and assuming uniform plane-wave motion along the cavity axis, the cavity admittance measured in the plane of the EPA-tip is given by the input admittance Y of a closed cylindrical tube of length L and cross-sectional area A (and hence, volume $V = AL$) as:

$$Y = j \frac{A}{\rho c} \tan(2\pi fL/c) \tag{2}$$

where f is frequency, $\rho$ is the density of air = $1.18\text{x}10^{-3}$ g/cm$^{3}$, and c is the speed of sound in air = $3.45\text{x}10^{4}$ cm/sec (c.f., Secs. 2.4 and 5.5 of Beranek, 1954). At frequencies well below the first resonance, $f \ll c/4L$, Eq. 1 can be approximated by:

$$\tilde{Y} = j \frac{2\pi fV}{\rho c^{2}} = j2\pi fC \tag{3}$$

where $\tilde{Y}$ is the admittance of an acoustic compliance $C = V/\rho c^{2}$. For all computations of the cavity admittances we have used Eq. 2; however, for f < 2 kHz, the differences between the values obtained from Eq. 2 and from Eq. 3 are small. For example, for the 1.0 and 2.0 cc cavities with L = 1.01 cm, the first resonance occurs at $f = c/4L \simeq 8.6$ kHz and $20\log_{10}|Y/\tilde{Y}|$ is 0.38 dB at 2 kHz (and 1.72 dB at 4 kHz). [For the 0.5 cc cavity with L = 0.51 cm, the first resonance occurs at ~17 kHz.] In Sec. II. E measurements in these three cavities are used to evaluate the Norton equivalent model referred to the plane of the EPA-tip.

## II. B. Further description of the electroacoustic-probe-assembly (EPA)

The volume velocity $U_t$ is derived from a dynamic earphone (Dyna Magnetics Devices D308). Although it would be acoustically advantageous (i.e., for maximum deliverable $U_t$ over the broadest frequency range) to position the volume velocity source near to the plane of admittance measurement, the size and mass of this earphone precluded mounting it on the earpiece unit. Instead, therefore, it is mounted on the headset assembly (described below) and the earphone output is coupled to the earpiece by a section of plastic tubing of ~5 cm in length and 0.90 mm inner diameter. Cotton was placed in this tube to reduce effects of tube/earphone resonances. The input voltage to the EPA, $V_i$, is applied to the earphone through an amplifier with a high-pass frequency response formed from a one-pole, RC filter with $f_{-3\ dB} \simeq 1300$ Hz. This filtering approximately "equalizes" the low-pass influence of the earpiece unit (described below).

To measure the sound pressure $P_t$, a miniature electret-condenser microphone (Thermo Electron 2340) is used. The microphone is mounted directly on the earpiece by fitting a short tubing section (0.5 cm in length and containing some cotton) over the microphone port. The microphone output $V_{pm}$ is fed to a preamplifier (designed and built in our laboratory) having adjustable voltage gain from 100 to 10,000 (40 to 80 dB) and having four filters, or "weightings", providing different degrees of low-frequency attenuation. These filters removed large-amplitude low-frequency acoustic-noise components present in the microphone output for the measurements in ears. Most measurements were made using a two-pole, high-pass RC filter with $f_{-3\ dB} = 250$ Hz (and a 12 dB/octave rejection rate at lower frequencies).

Static pressures are derived from the pressure system of a Grason-Stadler Otoadmittance Meter (GS-1720). The GS-1720 pressure output is connected to a control valve and then to one side of a differential pressure transducer (Statham Instruments PM6±1-350). The pressure tube is connected to the earpiece unit with a two-stage plastic-tubing connection including a section 120 cm in length with inner diameter 1.65 mm followed by a section 30 cm in length with inner diameter 0.90 mm. The other side of the pressure transducer is left open to ambient pressure. The pressure transducer is electrically interfaced to a sensitivity (gain) and offset control circuit which connects to a sensitive digital voltmeter (Data Precision 2440). The DC pressure indicating system is calibrated (a) by adjusting the offset for zero output when both sides of the pressure transducer are at ambient pressure, and (b) by adjusting the sensitivity, using a reference manometer, so that the voltmeter indicates the DC pressure (re ambient) directly in mm $H_2O$.[3]

The earpiece unit is built around the earpiece supplied with the GS-1720 Otoadmittance Meter. In order to reduce the internal admittance of the EPA, $Y_s$, and to insure that resonances in the earpiece channels and in the tubes and transducers connected to the earpiece are well damped and effectively "decoupled" from the load admittance $Y_\ell$, a narrow tube was inserted into each of the three earpiece channels. Polyethylene tubing (Intramedic, Clay Adams) of inner diameter 0.57 mm and outer diameter 0.96 mm was used; this outer diameter allowed the tubing to just pass through the earpiece channels. The tubes in the volume velocity and static pressure channels were 3 cm in length, just exceeding the length of the earpiece channels, and they were inserted so as to end flush with the surface of the earpiece tip. The tube in the microphone channel exceeded 3 cm by the length of the probe-tube extension (0 to 4 mm) determined for each test session (see Sec. II. E).

The earpiece is hermetically sealed in the earcanal using a silicone-fluid-filled inflatable-cuff supplied from Grason-Stadler (3 mm size). The cuff provides a comfortable seal and it facilitates the somewhat delicate placement of the earpiece, with the protruding extension, in the earcanal (see Sec. II. H).

A headset assembly for the EPA was built from a circumaural hearing protector (Mine Safety Appliances Noisefoe Mark IIM). On one side, the earcup was machined off leaving only that portion supporting the circumaural cushion. To obtain wider access to the ear, a segment of the lower portion of the cushion and cushion-plate was also removed. The earphone was cemented to the cushion plate; the tubing from the earphone to the earpiece flexibly supports the earpiece unit. A "press" to squeeze the inflatable-cuff fluid reservoir was also cemented to the cushion plate. The three connections to the EPA, including the earphone and microphone cables and the static-pressure tube, were all afixed at the top of the headband thereby permitting the subject to move freely without straining the positioned earpiece.

## II. C. Frequency response system for stimulus control and measurement

### (1) Measurement scheme and apparatus

A computer-controlled system was used for measuring and storing frequency responses. The input section of the system (see Fig. 2) provides for the generation of a gated sinusoid of programmable amplitude and frequency to be presented to the system under test. The complex amplitude of this steady-state signal is denoted $V_i(f)$, or simply $V_i$. The output signal from the test system, with fundamental component $V_o$, is presented to an amplifier of programmable gain $A_o$. The variable gain is used to reduce the dynamic range of the signals presented to the measurement section which followed. To determine $V_o$, the signal from the amplifier ($A_oV_o$) is multiplied against the main output, $V_m$, of the input oscillator and also against its quadrature component $V_q = V_m \measuredangle 90$ degrees. The multiplier outputs are low-pass filtered to obtain the DC components (i.e. the average values) of the product waveforms. These DC components, VM and VQ, are proportional to the amplitudes of the components of $A_oV_o$ which are in phase and in phase quadrature with respect to $V_m$:[4, 5]

$$\begin{aligned} VM &= |V_m||A_oV_o|\cos(\measuredangle A_o + \measuredangle V_o) \\ VQ &= |V_q||A_oV_o|\sin(\measuredangle A_o + \measuredangle V_o) \end{aligned} \tag{4}$$

The filter outputs are sampled (and digitized) at the computer wherein $V_o$ is calculated. Because $|V_m| = |V_q| = 1.0$ volt, $V_o$ is calculated as:

$$\begin{aligned} |V_o| &= (VM^2 + VQ^2)^{1/2} / |A_o| \\ V_o &= \tan^{-1}(VQ/VM) - \measuredangle A_o \end{aligned} \tag{5}$$

where $\measuredangle A_o$ is measured beforehand and $|A_o|$ is known by the computer.

Crucial to the accuracy in determining $V_o$ is the 90-degree phase difference between $V_m$ and $V_q$ and also the linearity and offset voltage of the multipliers. For the oscillator used (Krohn-Hite 3031R) the $\measuredangle(V_q/V_m)$ is specified at 90.0 ± 0.25 degrees. For the multipliers used (Analog-Devices 425K) nonlinearities are less than 0.1% and the offset voltages were adjustable such that the low-pass filter outputs measured at the computer were within ±1 bit of zero on the A/D converter.

Some of the equipment in Fig. 2 was designed and built in our laboratory. The low-pass filters used were 2-pole, passive RC-filters with $f_{-3\ dB}$ = 12 Hz (and a 12 dB/octave rejection rate at higher frequencies). The electronic switch is functionally similar to the Grason-Stadler 829E. The programmable attenuator ($A_i$) has 512 steps of 1/4-dB, i.e. 0 to (nominally) 127.75 dB of attenuation. The programmable amplifier has gain $|A_o|$ = 1 or 10, i.e. 0 or +20 dB. [The choice of $A_i$ and $A_o$ will be described later in the section.]

Additional equipment, not shown in Fig. 2, included a frequency counter to monitor the oscillator's frequency and an oscilloscope to monitor $V_i(t)$ and $V_o(t)$.

### (2) Sweep parameters

The frequency-response measurements were made as "sweeps" incorporating 71 tone-pulses. The oscillator frequency increased discretely from one tone-pulse to the next; the frequency during the first tone-pulse was programmed to 62.5 Hz and the frequency during the last tone-pulse was 8 kHz. The measurement frequencies were equispaced in log(f) with 10 frequencies per octave. Hence, the frequency increased by $2^{0.1} \approx 1.072$ from one tone pulse to the next. [The oscillator was programmable to a resolution of 1 Hz.]

The duration of each tone-pulse was either 600 or 800 msec. The sequence of events within each tone-pulse cycle is shown in Fig. 3. Prior to turning on the electronic switch, the output amplifier was set to $|A_o| = 1$. After the 20 msec rise-time for the tone-pulse, a 200 msec delay was allowed for settling of the test-system response and settling of the low-pass filter outputs. A "spot-check" of $|V_o|$ was then made. If $|V_o|^2$ exceeded 0.002 volt$^2$ (i.e., $|V_o|$ exceeded -27 dB re 1 volt, where 1 volt is the maximum measureable signal), then $|A_o|$ was left at 1 and the computer proceeded to sample VM and VQ. If $|V_o|^2$ was below 0.002 volt$^2$, then $|A_o|$ was switched to 10, and additional 200 msec settling-delay was included, and the sampling of VM and VQ began at 530 msec into the tone-pulse. Eight samples each of VM and VQ were then obtained over an interval of 250 msec, after which the electronic switch was turned off. The sampling of VM and VQ was alternated and occurred approximately equispaced in time over the 250 msec. The samples of VM and VQ were averaged by the computer.[6] Prior to the onset of the next tone-pulse, an interval of 600 msec elapsed during which some computations and storage of results occurred and the computer set the system for the next tone-pulse.

The duration of the sweeps varied somewhat because the individual tone-pulse duration was contingent on the size of $V_o$. For the measurements in ears, the sweep durations were usually near 85 sec (with all tone-pulse durations being 600 msec except for some frequencies between 4 and 8 kHz and occasionally one or two frequencies between 2 and 4 kHz).

The input attenuator $A_i$ was set in one of two modes. In one mode, sweeps were made at a constant $V_i$ level with $A_i$ selected at the start of the sweep and set to the same value for all tone-pulses of the sweep. In the second mode, $A_i$ was set at each frequency so as to control the magnitude of the sound-pressure at the EPA-tip. Further details on the setting of $A_i$ are given below when the two modes are applied to determine the acoustic characteristics of the EPA (see Sec. II. E).

### II. D. Data processing

The computer calculated the complex amplitude $V_o$ after each tone-pulse using the 8-sample average of VM and VQ and Eq. 5. $|V_o|$ was converted to dB re 1 volt. These measurements of $V_o$ were stored on disk-memory as lists of 71 number-pairs for each sweep. After each sweep, the measurements were displayed on a storage-oscilloscope and/or printed out (at selective frequencies). At the end of an experiment, the measurements were transferred to magnetic tape from which the data could be retrieved for further processing which was done "off-line". All of this processing, e.g. to obtain estimates of the eardrum admittance, was done within the computer system; hard-copy outputs were available in print and/or in graphical form using an XY-plotter interfaced to the computer system.

The computer system used to make and process the frequency-response measurements was either a Digital Equipment Corporation PDP-8L or PDP-12. All computer programs were written in a modified version of FOCAL (Herman, 1973).

## II. E. Acoustic characteristics and calibration of the EPA

An initial series of measurements was made to obtain information about the two EPA transfer functions $K_i(f)$ and $K_o(f)^7$ so that the sound pressure at the EPA-tip, $P_t(f)$, could be controlled and calculated. For these measurements, the inflatable cuff was removed and the EPA-earpiece was hermetically coupled (using plastic tubing sections) to a calibrated reference microphone system (Bruel & Kjaer parts 4136, UA 0035, 2619, and 2604). The (cylindrical) cavity between the EPA-tip and the reference microphone was 5 to 10 mm in length and 1 $cm^2$ in cross-sectional area. With this configuration a sweep was made with constant input attenuation $A_i$ [i.e. constant $|V_i(f)|$] and the output from the reference microphone system was measured. Making use of the known sensitivity of the reference microphone system and assuming the EPA-earphone output is linearly related to input voltage $V_i(f)$, this measurement was converted to an estimate of the sound-pressure level that would result for maximum input voltage applied to the EPA earphone, see Fig. 4 ( ■ ).

This computed response was then used as a reference from which sweeps at an approximately constant sound-pressure level were made. Specifically, a desired sound-pressure level was "requested", denoted by $P_{REQ}$, and at each measurement frequency in the sweep the computer set the input attenuation $A_i$ to the difference in dB between the reference (i.e. "available") sound-pressure level and the desired $P_{REQ}$. Figure 4 ( ▲ ) shows an example of the sound-pressure level developed for a sweep when $P_{REQ}$ = 75 dB SPL. A second sweep was also performed at this same $P_{REQ}$ and the output from the EPA probe-tube microphone system, $V_o(f)$, was measured. The measurements from these two sweeps, along with the stored frequency response and gain of the microphone preamplifier, provided for an absolute calibration of the EPA probe-tube/microphone combination, $K_{pm}(f) = P_t(f)/V_{pm}(f)$.

All further measurements were performed using control of $V_i(f)$ based on the reference response described above. We call this sweep mode a "requested sound-pressure level", although when $Y_t(f)$ differs from that occurring in the reference coupler cavity, $P_t(f)$ differs from $P_{REQ}$ and it is precisely such differences [as indicated through $V_{pm}(f)$ changes] that will be utilized to determine unknown $Y_\ell(f)$'s. Nevertheless, we use the "$P_{REQ}$" description because (as will be seen in Sec. II. H) $|P_t(f)|$ developed within the earcanal is within a few dB of $P_{REQ}$.

The next series of tests was directed toward determining $Y_s(f)$ and demonstrating the validity of the Norton equivalent model for the EPA. With the EPA joined to each of the three precision test-cavities, of volumes 0.5, 1.0, and 2.0 cc, a sweep was made with the same $P_{REQ}$ [i.e. the same $V_i(f)$] and the EPA output voltage $V_o(f)$ was measured. The values of $V_o(f)$ were converted to estimates of $V_{pm}(f)$. Results for $|V_{pm}(f)|$ are shown in Fig. 5 for a case where the microphone probe-tube extended 3.5 mm beyond the EPA-tip.

The measurements of $|V_{pm}(f)|$ in the 0.5 and 2.0 cc cavities have a frequency dependence that is similar to those for the 1.0 cc cavity for $f < 5$ kHz. Between about 125 and 2000 Hz, $|V_{pm}(f)|$ appears to change roughly in proportion to the inverse change in cavity admittance. Specifically, reductions by 6 dB (a factor of 2) in cavity admittance which occur in changing from the 2.0 to the 1.0 cc cavity, and also from the 1.0 to the 0.5 cc cavity, result in increases of ~6 dB in $|V_{pm}(f)|$. These magnitude results indicate that $|Y_s(f)|$ is small compared to $|Y_{1\ cc}(f)|$ for $125 < f < 2000$ Hz.

Fig. 5 also includes results from two additional sweeps with the same $P_{REQ}$ but (1) when the EPA-earpiece was removed from the cavities and was left open and directed away from nearby reflecting surfaces, a "free radiation" condition, and (2) when a tapered wire plug was pressed into the tip of the microphone probe-tube and the earpiece was reinserted into the 1.0 cc cavity. The first of these measurements constitutes an attempt to present a maximum admittance to the EPA-tip. The second measurement constitutes a test for artifacts, i.e. contributions to $V_{pm}(f)$ not directly resulting from $P_t(f)$.

The measurements of $|V_{pm}(f)|$ with the plugged probe-tip (PPT) and the "free radiation" (FR) conditions are substantially below $|V_{pm}(f)|$ obtained in the test cavities for $f \lesssim 1.4$ kHz. Above this frequency, $|V_{pm}(f;PPT)|$ and $|V_{pm}(f;FR)|$ increase and approach the values of $|V_{pm}(f)|$ obtained in the test cavities. Furthermore, because $|V_{pm}(f)|$ is comparable for both the PPT and FR conditions, and because an additional measurement (not shown) with no input signal [i.e. $V_i(f) = 0$] results in $|V_{pm}(f)|$ in the range of -15 to 0 dB re 1 uv for $f > 200$ Hz, we conclude that the indicated PPT and FR results above ~1.4 kHz represent a "true" artifact. Above ~6 kHz, this artifact dominates the measurements; $|V_{pm}(f)|$ is nearly independent of the load condition at the EPA-earpiece tip. Thus, admittance measurements in this frequency range will not be possible with the system in this configuration. At 4 kHz, however, the artifact is ~18 dB below $|V_{pm}(f)|$ obtained in the 1.0 cc cavity and the measurements can provide an accurate indication of admittance.

The measurements of $V_{pm}(f)$ from the test cavities (magnitude and angle) were used to estimate $Y_s(f)$. Specifically, Eq. 2 is used to compute the theoretical admittances of the cavities, these admittances and the associated $V_{pm}(f)$ [and $V_i(f)$] measurements were then taken pairwise and used with Eq. 1 to obtain an estimate of $Y_s(f)$. Three such estimates of $Y_s(f)$ were obtained (though they are not independent), see Fig. 6.

To the extent that the EPA can be described by a Norton equivalent model and that the cavity admittances are accurately given by Eq. 2, then these three estimates of $Y_s(f)$ should be identical. The results in Fig. 6 indicate that the three estimates of $Y_s(f)$ are similar at low frequencies but disparate for $f \gtrsim 500$ Hz. At low frequencies, $\measuredangle Y_s(f)$ is near zero, and hence, $Y_s(f)$ is dominated by its real component $G_s(f)$ [i.e. $G >> B$, where $Y = G + jB$, G = conductance, and B = susceptance]. Furthermore, an approximately constant value of $G_s(f) = 0.16$ mS (-16 dB re 1 mS) is indicated for $62 \lesssim f \lesssim 125$ Hz; this $|Y_s|$ is comparable to $|Y_{0.5\ cc}(f)| = |Y_{1\ cc}(f)|$ -6 dB in this frequency range. In considering the discrepancies between the three estimates of $Y_s(f)$ above 500 Hz, two important factors are involved. (1) When $|Y_s(f)|$ is much smaller than the admittance magnitude of the test cavities, estimations of $Y_s(f)$ are inaccurate because of high sensitivity to noise in the measurements and to discrepancies between the "actual" admittances of the cavities and the values given by Eq. 2. (2) When $Y_s(f)$ is small, errors in its estimation are less important than when it is comparable to the admittances being measured. Thus, the differences in the estimates of $Y_s(f)$ for

$0.5 \lesssim f \lesssim 4$ kHz are relatively unimportant because $|Y_s(f)|$ is small in this region. Above 4 kHz, however, the estimates of $|Y_s(f)|$ approach and then exceed the theoretical admittance magnitude of the 1.0 cc cavity; the estimates of $\measuredangle Y_s(f)$ are not plotted because they appear to be random over the range of ± 180 degrees. In this region, therefore, the discrepancies between the different estimates of $Y_s(f)$ indicate that the measurements in the three test cavities are not consistent with the theoretically specified cavity admittances and a single Norton equivalent model for the EPA. These discrepancies arise principally because of the artifact noted in the discussion of Figure 5.

The extent to which the three estimates of $Y_s(f)$ are consistent with a single Norton equivalent model for the EPA can be quantitatively considered as follows. An estimate of the "true" $Y_s(f)$ was obtained by averaging, in dB and degrees, the three $Y_s(f)$ estimates shown in Fig. 6. This function, the 1.0 cc cavity measurements of $V_{pm}(f)$ and $V_i(f)$, and the theoretical admittance of the 1.0 cc cavity were then used as the "reference" characteristics for the EPA. The measurements in the 0.5 and 2.0 cc cavities were treated as data and the load admittances implied by each of these measurements were obtained using Eq. 1. The solid lines of Fig. 7 show the "errors", i.e. deviations, between these measured cavity admittances and the theoretical admittance values for the 0.5 and 2.0 cc cavities. Between 125 and 2000 Hz, these errors are very small, being within ± 0.2 dB and ± 1.0 degree. At lower and higher frequencies, the errors increase but remain within ± 1.0 dB and ± 10 degrees for $0.62 \lesssim f \lesssim 4$ kHz. Between 4 and ~5.6 kHz, the errors are within ± 2.0 dB and ± 20 degrees; at higher frequencies the errors increase rapidly. These results give an indication of the accuracy of admittance measurements with this system.

Figure 7 also shows, for low and intermediate frequencies, the extent to which the "raw" $V_{pm}(f)$ measurements in the 0.5 and 2.0 cc cavities divided by those in the 1.0 cc cavity deviate, respectively, from the theoretical cavity admittance ratios $Y_{1\ cc}(f)/Y_{0.5\ cc}(f)$ and $Y_{1\ cc}(f)/Y_{2\ cc}(f)$. [$V_i(f)$ was the same for the three cavities.] These deviations represent the errors that would occur in computing the 0.5 and 2.0 cc cavity admittances from the $V_{pm}(f)$ measurements if $Y_s(f)$ was assumed to be zero. Of particular note are the angle deviations at low frequencies which result because of the influence of the sizeable $G_s(f)$ in this region. These angle deviations are important because the angle of the admittance at the eardrum approaches +90 degrees at low frequencies and hence, estimation of the real part of the eardrum admittance at low frequencies is very sensitive to the accuracy of the $G_s(f)$ estimates.

The measurements and computations shown in Figs. 5, 6, and 7 are for an EPA incorporating a microphone probe-tube extension of 3.5 mm. In the measurements on the subjects, the extension lengths that were used were determined in each experimental session and varied from 0 to 4.0 mm (see Sec. II. H). The effect of reducing the probe-tube extension from 3.5 to 0 mm (wherein the tip of the probe tube is flush with the face of the earpiece) is to reduce the "useable" upper frequency limit for admittance measurement by ~2/3 of an octave, from 4.0 to 2.5 kHz. In particular, the "free radiation" measurement, $V_{pm}(f;FR)$, increases re that which occurs with a probe-tube extension, and $|V_{pm}(f;FR)|$ and $|V_{pm}(f)|$ obtained in the three precision cavities all intersect at ~4 kHz. Thus, near 4 kHz, large changes in $Y_\ell(f)$ do not result in corresponding changes in $V_{pm}(f)$; this "insensitivity" to $Y_\ell(f)$ occurs because (as noted in Sec. II. A) the measurement of $P_t(f)$ in the immediate vicinity of the volume velocity inlet is "contaminated" by modes other than that related to $Y_\ell(f)$. Additionally, the three estimates of $|Y_s(f)|$ intersect $|Y_{1\ cc}(f)|$ at ~4 kHz, and the errors in computing the admittances of the 0.5 and 2.0 cc cavities reach ±1.0 dB and ±10 degrees at ~2.5 kHz.

In summary, we have shown that a Norton equivalent model, and particularly a source admittance $Y_s(f)$, can be determined with sufficient accuracy for an EPA with a 3.5 mm probe-tube extension such that the measurements of $V_{pm}(f)/V_i(f)$ obtained in three precision cavities are consistent with the theoretical admittances of these cavities to a precision given by the solid lines of Fig. 7. For an EPA with no probe-tube extension, results similar to those in Fig. 7 are obtained, but for a 2/3-octave reduction in the high-frequency range for which "errors" begin to occur.

## II. F. A supplementary admittance-measuring test

A supplementary experiment was conducted in which the admittance of two narrow tubes was measured and compared to theoretical admittance values. This experiment tested the EPA, and its Norton description, in measuring a load admittance with both real and imaginary components (as compared to the precision cavity tests which involved only loads with $\measuredangle Y_\ell(f) = +90$ degrees).

A 0.5 cc cavity was machined that was similar to the precision 0.5 cc cavity described above, except for two channels that were drilled radially (and diametrically opposed) and midway along the length of the cavity. Into these channels fit tubes such that the tube openings were flush with the wall of the cavity. Each tube was of inner diameter $D = 0.211$ cm (cross-sectional area $A = 0.0349$ cm$^2$) and length $L = 43.1$ cm; the tubes were closed at their distal ends.

For tubes of this diameter, significant viscous and thermal losses occur in the boundary layer of the sound wave near the walls of the tube (c.f., Secs. 344-350 of Rayleigh, 1945; Sec. 6.4 of Morse and Ingard, 1968). An estimate of these losses (Eq. 5.4 of Beranek, 1949; Eqs. 15-20 of Sec. 350 in Rayleigh, 1945) in conjunction with the theory of dissipative transmission lines (c.f., Sec. 7.4 of Beranek, 1949) provides an estimate of the input admittance of such a tube as:

$$Y_{tu} = \frac{A}{\rho c} (1 - j\alpha/\beta) \tanh[(\alpha + j\beta)L] \qquad (6)$$

where, $\alpha = 2\xi\sqrt{\pi f}/cD$

$\beta = 2\pi f/c'$

$c' = c/[1 + \xi/D\sqrt{\pi f}]$ = speed of sound in the tube

$\xi = 0.571\ \mathrm{cm/sec}^{1/2}$, $\xi$ accounts for the viscous and thermal losses in the air near the walls of the tube[8]

$f$ = frequency in Hz

$\rho = 1.18\mathrm{x}10^{-3}\ \mathrm{g/cm}^3$ = density of air

$c = 3.45\mathrm{x}10^4$ cm/sec = speed of sound

L, D, and A = length, diameter, and cross-sectional area of the tube.

Because the 43.1 cm tube has resonances spaced at roughly 400-Hz intervals [with a first resonance at $f = c'(f)/4L \approx 175$ Hz], the frequency table of the sweep system was modified from its usual logarithmic scale to a linear scale in which the measurement frequency incremented in 40-Hz steps from 100 to 3900 Hz (96 measurement frequencies). Using this modified sweep system, an EPA with a 2.0 mm probe-tube extension was calibrated, and an average $Y_s(f)$ determined, as described in Sec. II. E. A sweep was then made with the EPA joined to the 0.5 cc cavity loaded with the two tubes; $V_{pm}(f;0.5\ cc + 2tu)$ was measured using the same $V_i(f)$ as was used for the precision cavity sweeps. The load admittance at the EPA tip, $Y_\ell(f)$, implied by this measurement was obtained using Eq. 1 (and the 1.0 cc cavity as the "reference"). An estimate of the admittance of the two tubes, $Y_{2tu}(f)$, followed assuming that $Y_{2tu}(f)$ was in parallel with the theoretical admittance of the 0.5 cc cavity, $Y_{0.5\ cc}(f)$; i.e., $Y_{2tu}(f) = Y_\ell(f) - Y_{0.5\ cc}(f)$.

In Fig. 8, the measured $Y_{2tu}(f)$ is shown along with the theoretical admittance given by Eq. 6 (multiplied by 2, and computed in 20-Hz increments). The agreement between the measured and theoretical values of $Y_{2tu}(f)$ is considered good, although some differences exist. Some of these differences appear systematic: (1) Below ~2 kHz, there is tendency (a) at the maxima in $|Y_{2tu}(f)|$, for the measured (magnitude) values to fall below the theoretical values, and (b) at the minima of $\measuredangle Y_{2tu}(f)$, for the measured (angle) values to fall above the theoretical values. (2) Throughout the entire frequency range tested, the measured functions, for both magnitude and angle, are shifted slightly toward higher frequencies re the theoretical functions. These systematic differences may, however, reflect deviations between the "actual" tube admittance and the theoretical admittance given by Eq. 6.[9] Thus, we reiterate that the agreement in Fig. 8 is considered good; we conclude that the Norton model for the EPA appears valid for measuring load admittances with any phase angle, even though the model is determined solely from tests on load admittances having $\measuredangle Y_{\ell}(f) = +90$ degrees.

## II. G. Subjects

Four males served as subjects: three undergraduates aged 21 to 22 and the author aged 28. All reported a negative history of otic disease. The subjects were referred to an otologist for examination; all were judged to have tympanic membranes of normal appearance and mobility, and pure-tone thresholds were within normal limits, i.e. within 15 dB of ISO-1964 recommended zero from 0.25 to 4 kHz.

The subjects are denoted S1 to S4; the author was S1.

## II. H. Experimental protocol

All experiments were conducted with the subject seated upright within a double-walled soundproof room. The EPA system was also located in the soundproof room; the computer-controlled frequency-response system was outside the soundproof room. Before each session, the input section of the EPA was calibrated (see Sec. II. E for details).

Each subject's right ear was used. At the start of each session, the experimenter inspected the earcanal with the aid of an otoscope; cerumen on the wall of the canal was removed with an alcohol wetted swab.

The length of microphone probe-tube extension was then determined as follows: The microphone was removed from the EPA and an excess length of probe-tubing was passed into the microphone channel so that the tip of the tubing was flush with the tip of the EPA. The EPA was then positioned in the earcanal and the cuff was inflated. The tubing in the microphone channel was then passed inward (in steps $< 1$ mm) until the subject felt a gentle "poke" to the wall of the earcanal or until a maximum allowed extension of 4.0 mm (beyond the EPA tip) was reached. If poking occurred with an extension of less than 3.0 mm, the EPA was removed and then repositioned in an attempt to permit use of a longer probe-extension. That is, the experimenter attempted to obtain a "better" direction and orientation for the EPA insertion. As many as four attempts were made to obtain a position permitting an extension of 3.0 mm.

The EPA was then removed from the subject's ear and the microphone was re-attached with a probe-extension cut to the above determined length. The EPA was then reinserted into the subject's earcanal, the cuff was inflated. The quality of the hermetic seal was judged adequate if static pressures of $P_{DC}$ = +40 and -40 cm $H_2O$ were maintained with essentially no observable decay (i.e., a pressure drop of $<$ 0.5 cm $H_2O$ over ~15 sec). On some occasions seals of this quality were obtained on the first EPA insertion; sometimes a few "attempts" were needed. If the seal was inadequate, the cuff was deflated, the EPA was repositioned slightly (usually further inward), and the cuff was re-inflated. In addition to requiring an adequate hermetic seal, it was required that the probe-extension not be in contact with the earcanal wall. Although the probe-extension was cut to the length for which "poking" had previously occurred, the inward passing of the tubing (in the determination of the extension length) always involved movement of the EPA, both inward and rocking. Hence, the EPA with that pre-measured probe-extension could usually be sealed in the earcanal with the probe-extension not contacting the earcanal wall. Subject S4, however, had an earcanal with sharp bends, and no extension was usable. While an allowed extension of ~ 2 mm was measured for a given EPA position, insertion of the EPA involved pushing the EPA beyond its eventual "seated" position, and the insertion itself always resulted in the probe-extension poking the earcanal wall.

The probe-extensions used were 3.5 to 4.0 mm for S1 and S2, 2.0 to 2.5 mm for S3, and zero for S4.

With the EPA hermetically sealed in the earcanal, the admittance measurements were begun. The initial measurements always consisted of three sweeps, one each with earcanal static pressure $P_{DC} = 0$, +40, and -40 cm $H_2O$, and with a moderate sound-pressure level requested, $60 \leq P_{REQ} \leq 75$ dB SPL (based on the stored calibration of the input section of the EPA). Next, measurements were made: (a) with $P_{DC} = 0$, and different values of $P_{REQ}$, (b) with $P_{DC} = 0$, $60 \leq P_{REQ} \leq 75$, and intense noise presented to the contralateral ear to elicit middle-ear-muscle contractions, (c) with $60 \leq P_{REQ} \leq 75$, and different values of $P_{DC}$ between ± 40 cm $H_2O$, and (d) in the nominal "resting" condition with $P_{DC} = 0$, $60 \leq P_{REQ} \leq 75$, and no contralateral noise. To conclude the in-ear measurements, sweeps were again made with $P_{DC} = 0$, +40, and -40 cm $H_2O$, and $60 \leq P_{REQ} \leq 75$.[10] [The measurements from (a) are discussed in the next section; the results from (b) are reported ahead (see Part 2); the results from (c) will appear at a later date; and, the results from (d) along with those from the initial and concluding sweeps form the present results (Sec. III).] The EPA was then removed from the earcanal, and the earpiece was inspected. No moisture or cerumen was observed on the earpiece tip or the probe-extension, although some cerumen was sometimes present on the inflatable cuff. The cuff was then removed from the earpiece, and measurements were made in the three precision cavities of volumes 0.5, 1.0, and 2.0 cc. [These measurements were used to calibrate the EPA for determining the admittances implied by the in-ear sweeps.]

The number of in-ear sweeps in the test sessions ranged from 12 to 36. The time the EPA was in a subject's ear ranged from about 20 to 75 minutes.

During the sweep measurements the subject was instructed to remain stationary and not to swallow. In the intervals between successive sweeps (about 20 to 90 sec), the subject relaxed while the experimenter set the computer system for the next sweep and made any (needed) adjustments in the earcanal static pressure $P_{DC}$. In an attempt to insure near-zero static pressure within the middle-ear cavities, the subject was instructed to swallow prior to most sweeps and while $P_{DC}$ was at zero.

## II. I. Selection of sound-pressure level

A compromise was involved in the selection of the ("probe-tone") sound-pressure level to use in the eardrum-admittance measurements. A high sound-pressure level gives a good signal-to-noise ratio, but too high a level may yield different values of eardrum admittance (re that which is measured at lower levels) because of (a) acoustically elicited (ipsilateral) contractions of the middle-ear muscles, and/or (b) passive nonlinearities in the mechanics of motion of the middle-ear system.

A test was performed in which the proportionality between the (fundamental component of the) EPA-microphone output, $V_{pm}(f)$, and the input voltage to the EPA, $V_i(f)$, was examined. Specifically, with an EPA sealed in the subject's earcanal, $V_{pm}(f)$ was measured for successive sweeps at different values of the "requested" sound-pressure level $P_{REQ}$ [i.e. different $V_i(f)$'s]. The measurements of $V_{pm}(f)$ were converted [using the stored response of $V_{pm}(f)/P_t(f)$] to estimates of the sound-pressure in the earcanal at the EPA tip, $P_e(f) = P_t(f)$. Typical results are shown in Fig. 9.

To facilitate consideration of the proportionality between $P_{REQ}$ and $|P_e(f)|$ [and hence, between $|V_i(f)|$ and $|V_{pm}(f)|$], the values of $|P_e(f)|$ for $P_{REQ}$ = 75 dB SPL are shown as measured (—○—) and also shifted vertically in 10 dB steps (plotted as dots). These shifted responses generally superimpose on the results obtained for the other values of $P_{REQ}$ except (a) for instances of measurement noise, and (b) for $f > 3$ kHz, $|P_e(f)|$ for $P_{REQ}$ = 85 dB SPL falls below that for

$P_{REQ}$ = 75 dB SPL shifted up to 10 dB because of a limitation on the "available" input voltage to the EPA.

The angle measurements corresponding to the magnitude results in Fig. 9 were examined in terms of the functions $\measuredangle[P_e(f;P_{REQ})/P_e(f;P_{REQ} = 75)]$. There were no systematic deviations from zero for these functions, though significant measurement noise is evident in the results for $P_{REQ} \lesssim 55$ dB SPL.

The above results indicate that ("probe-tone") sound-pressure levels (at least) as high as ~85 dB SPL may be used in measuring eardrum admittance with no detectable influence, <u>in the present measuring scheme</u>, of ipsilateral acoustic-reflex action or other nonlinearities. We generally used values of $|P_e(f)| \lesssim 75$ dB SPL (and always $\lesssim 80$ dB SPL) for eardrum-admittance measurement. For adequate signal-to-noise ratio, we used $|P_e(f)| \gtrsim 60$ dB SPL.

Finally, Fig. 9 demonstrates that the $P_{REQ}$ mode of input voltage control does provide approximate control of $|P_e(f)|$, i.e. $|P_e(f)| = P_{REQ} \pm 5$ dB.

## III. Results

### III. A. Admittance in the plane of the EPA-tip, $Y_e(f)$

The in-ear sweeps from each session were converted to estimates of the admittance of the ear in the plane of the EPA-tip. This admittance, denoted $Y_e(f)$, was calculated from Eq. 1 with $Y_\ell(f) = Y_e(f)$ and $Y_s(f)$ determined from the precision-cavity tests and with the 1.0 cc cavity as the "reference" condition.

For typical sessions on two subjects, Fig. 10 shows results for $Y_e(f)$ for two series of three sweeps performed at the "start" and "end" of the sessions. Each series includes measurements with static pressure $P_{DC} = 0$, +40, and -40 cm $H_2O$. Comparison of the results at the "start" and "end" shows the good stability of the measurements.

At low frequencies and for all three conditions of $P_{DC}$[11], $Y_e(f)$ can be approximated by an equivalent compliance in that $\measuredangle Y_e(f)$ is near +90 degrees and $|Y_e(f)|$ increases approximately 6 dB per octave [as indicated by the near parallel of $|Y_e(f)|$ and $|Y_{1\ cc}(f)|$]. For the positive and negative pressures these low-frequency equivalent compliances are about equal and they are smaller than the compliance measured at ambient pressure.

We have computed low-frequency compliances in terms of equivalent volumes for the $Y_e(f)$ data by averaging the dB difference between $|Y_e(f)|$ and $|Y_{1\ cc}(f)|$ for $100 \leq f \leq 300$ Hz, dividing this number by 20, and then taking the antilogarithm. The compliance values obtained with positive and negative pressure ranged from 0.61 to 0.92 cc over all subjects and sessions with an overall average of 0.76 cc. For ambient pressure, the range was 1.04 to 1.46 cc with an average of 1.24 cc.

With the positive pressure, $Y_e(f)$ is fairly well described as a compliance up to ~2 kHz. With the negative pressure, however, $|Y_e(f)|$ above ~700 Hz is a few dB below that measured with positive pressure and for $0.7 \lesssim f \lesssim 2$ kHz the equivalent compliance is decreased re that indicated for lower frequencies. At ambient pressure, $Y_e(f)$ departs from an equivalent compliance for frequencies above a few hundred Hz. With increasing frequency $\measuredangle Y_e(f)$ decreases and reaches a (local) minimum of +25 to +50 degrees between 1 and 1.5 kHz. Also, the growth of $|Y_e(f)|$ decreases near the $\measuredangle Y_e(f)$ minimum, and $|Y_e(f)|$ approximates that measured with the negative pressure. These features were observed in all sessions on all four subjects.

## III. B. Transformation of $Y_e(f)$ to the admittance at the eardrum, $Y_d(f)$

To obtain the admittance at the eardrum $Y_d(f)$ from the admittance $Y_e(f)$, the space between the eardrum and the EPA-tip must be characterized. We have represented this "residual earcanal space" as a rigid-walled tube of length $L_C$ and uniform cross-sectional area $A_C$. As a result, $Y_d(f)$ and $Y_e(f)$ are related by a lossless transmission-line transformation (c.f., Sec. 8.7 of Kinsler and Frey, 1962):

$$Y_d(f) = \frac{Y_e(f) - j\,\dfrac{A_C}{\rho c}\tan(2\pi f L_C/c)}{1 - j\,\dfrac{\rho c Y_e(f)}{A_C}\tan(2\pi f L_C/c)} \qquad (7)$$

in which the two parameters $L_C$ and $A_C$ must be determined. At low frequencies, $\tan(2\pi f L_C/c) \rightarrow 2\pi f L_C/c$ and $|Y_e(f)|/(A_C/\rho c) << 1$ [for $A_C$ = 0.48 $\text{cm}^2$, $A_C/\rho c$ = 11.7 mS (+21.4 dB re 1 mS)], and Eq. 7 can be approximated as:

$$Y_d(f) = Y_e(f) - j\,\frac{2\pi f V_C}{\rho c^2} \qquad (8)$$

where $V_C = A_C L_C$ = the volume of the residual earcanal space. Thus at low frequencies, (a) $Y_e(f)$ is the sum of $Y_d(f)$ and the admittance of the residual earcanal space, and (b) the admittance of the residual earcanal space is that of a compliance proportional to its volume $V_C$.

For all subjects we have used an "effective" earcanal area $A_C$ = 0.48 $\text{cm}^2$. This value was chosen by considering earcanal dimensions reported in Zwislocki (1970, Sec. 3.2) and Shaw (1974b, Sec. D.2). In particular, the value $A_C = 0.48\ \mathrm{cm}^2$ was chosen to be consistent with earcanal dimensions from the eardrum to the entrance of the earcanal of volume 1.1 cc and length 2.3 cm (for adult males). Because $A_C$ is assumed to be constant, to apply Eq. 7 the parameter $L_C$ (or $V_C$) must be determined for each session.

To determine $V_C$ we have used the low-frequency equivalent compliance values calculated from $|Y_e(f)|$ at ambient pressure and at the average of $P_{DC} = +40$ and $-40$ cm $H_2O$. It was noted (Sec. III. A) that when $P_{DC}$ changes from zero to +40 or -40 cm $H_2O$, a reduction in equivalent compliance occurs. We assume that all of this compliance reduction is attributable to a reduction of the compliance at the eardrum $V_d$, i.e. that $V_C$ is unchanged:

$$V_C + V_d = V_e(P_{DC} = 0) \tag{9a}$$

$$V_C + \kappa V_d = V_e(|P_{DC}| = 40) \tag{9b}$$

where (a) $V_e$ denotes equivalent volume calculated from $|Y_e(f)|$ at low frequencies, $V_e(P_{DC} = 0)$ being the compliance obtained in the ambient pressure condition and $V_e(|P_{DC}| = 40)$ being the compliance averaged for the pressure conditions of +40 and -40 cm $H_2O$, and (b) $\kappa < 1$ represents the effect of $|P_{DC}| = 40$ cm $H_2O$ on the (low-frequency) compliance at the eardrum.[12] Thus, $V_C$ (and $V_d$) can be determined from the measured values of $V_e$ if the parameter $\kappa$ can be determined.

In the clinical use of "tympanometry" (c.f. Sec. III of Northern, 1976) it is routinely assumed that large static pressure, $|P_{DC}| \geqslant 20$ cm $H_2O$, causes a fixation of the middle-ear system. In terms of Eq. 9, this corresponds to $\kappa = 0$, and therefore, $V_C = V_e(|P_{DC}| = 40)$.[13] However, we have not assumed $\kappa = 0$ for two reasons. First, it is clear that $\kappa$ exceeds zero because the admittance measuring tones are audible to the subject. Increasing $|P_{DC}|$ from zero to 40 cm $H_2O$ does reduce the loudness of the tones, but the reductions are moderate. Second, we will demonstrate below that assuming $\kappa = 0$ leads to estimates of $Y_d(f)$ which are difficult to reconcile with reported measurements of the sound-pressure transformation of the earcanal, i.e. the sound-pressure magnitude at the eardrum divided by the sound-pressure magnitude at the entrance to the earcanal.

The problem is to decide what value of $\kappa \neq 0$ is appropriate. To make this decision we have used measures of middle-ear transmission change for $-40 \leqslant P_{DC} \leqslant +40$ cm $H_2O$[14] showing that (a) for $P_{DC} = +40$ and $-40$ cm $H_2O$, transmission reductions are nearly independent of frequency below 630 Hz, (b) below 630 Hz, the transmission reductions increase with $|P_{DC}|$ almost symmetrically for positive and negative pressure, and (c) the transmission change averaged for $P_{DC} = +40$ and $-40$ cm $H_2O$ and averaged for f = 400 and 500 Hz (denoted $\Delta|H|$) ranged from -14.5 to -20 dB for the four subjects, with an average of -16.4 dB.

We have assumed that the effects of $|P_{DC}| = 40$ cm $H_2O$ on the low-frequency eardrum admittance and middle-ear transmission are proportional for each subject so that $\kappa$ is given by:

$$20\log_{10}\kappa = \Delta|H| + \delta \tag{10}$$

where $\Delta|H|$ is in dB and $\delta$ is a constant (in dB) which is assumed to be the same for the four subjects.

To select a value for $\delta$ we have made use of other measurements of average earcanal properties. Specifically, the magnitude of the sound-pressure transformation between the eardrum and the entrance to the earcanal, $|P_d(f)/P_{ec}(f)|$, has been measured in a number of studies (Wiener and Ross, 1946; Yamaguchi and Sushi, 1956; Jahn, 1960; Djupesland and Zwislocki, 1973, see also Zwislocki, 1970). While substantial intersubject variations are reported (Wiener and Ross, 1946; Djupesland and Zwislocki, 1973) for $|P_d(f)/P_{ec}(f)|$, the averaged measurements from the cited studies are in relatively close agreement. Shaw (1974a, 1974b) has reviewed these measurements in detail; he suggests that ambiguities in the definitions of measurement positions are probably responsible for some of the inter-study differences which do exist. Furthermore, Shaw (1974a, Fig. 2, also 1974b, Fig. 13) has suggested representative values for $|P_d(f)/P_{ec}(f)|$ by fitting the mean results of Wiener and Ross (1946) and Djupesland and Zwislocki (1973). With our assumption that the earcanal can be modeled as a rigid-walled tube, this sound-pressure transformation is theoretically related to $Y_d(f)$:

$$\frac{P_d(f)}{P_{ec}(f)} = \frac{1}{\cos(2\pi f L_{EC}/c) + j\,\dfrac{\rho c Y_d(f)}{A_{EC}}\,\sin(2\pi f L_{EC}/c)} \tag{11}$$

where $A_{EC}$ = earcanal area = $A_C = 0.48\ cm^2$ and $L_{EC}$ = earcanal length = 2.3 cm (as noted earlier in this section). *Since measurements of the magnitude of this ratio are in close agreement (on the average), we investigated whether a particular value of $\delta$ could be chosen such that the group-average $Y_d(f)$ obtained for our four subjects would lead to estimates of $|P_d(f)/P_{ec}(f)|$ (calculated from Eq. 11) which would be consistent with the representative values given by Shaw (1974a).* After some trial and error, the value of $\delta = +5$ dB was found to satisfy this condition. For the average $\Delta|H| \simeq -16$ dB, this corresponds to $20\log_{10}\kappa = -11$ dB and $\kappa = 0.28$, i.e. $|P_{DC}| = 40$ cm $H_2O$ reduces the compliance at the eardrum to 28% of its value at ambient pressure.

Fig. 11 demonstrates the influence of different choices of $\delta$ on (a) $|P_d(f)/P_{ec}(f)|$ and the magnitude and angle of $Y_d(f)$, and (b) the real and imaginary components of the eardrum impedance, $Z_d(f) = R_d(f) + jX_d(f) = 1/Y_d(f)$. The results are not explicitly shown in terms of the parameter $\delta$, but in terms of the resulting variations in the estimate of $V_C$. These results were obtained as follows: With $\delta = +5$ dB and using each subject's $\kappa$ given by Eq. 10, $V_C$ for each session was calculated using Eq. 9. Estimates of $Y_d(f)$ were then obtained (for each sweep) using Eq. 7 and $A_C = V_C/L_C = 0.48\ cm^2$. A group-average[15] $Y_d(f)$ (indicated —●— in Fig. 11) and a group-average $V_C = 0.57$ cc were calculated. Using this $V_C$, $Y_d(f)$, and Eq. 7, an "average" $Y_e(f)$ was calculated. The $Y_d(f)$'s corresponding to different choices of $\delta$ (and $V_C$) were calculated from this $Y_e(f)$. This $Y_e(f)$ gives a value for $V_e(P_{DC}=0) = 1.24$ cc. The group-average $V_e(|P_{DC}|=40)$ was also calculated, its value was 0.76 cc. Using these $V_e$ values and Eq. 9,

$V_C$'s corresponding to different κ's were calculated; using the above determined $Y_e(f)$ and Eq. 7, corresponding estimates of $Y_d(f)$ were then calculated.

The results for three values of δ other than +5 dB are included in Fig. 11. The results for $V_C = 0.76$ cc correspond to the "limiting-case" of $V_e(|P_{DC}| = 40) = V_C$, i.e. $\delta = -\infty$ dB and $\kappa = 0$. The results for $V_C = 0.66$ cc correspond to assuming that the reduction in the low-frequency equivalent compliance at the eardrum caused by $|P_{DC}| =$ 40 cm $H_2O$ is equal to the low-frequency reduction in middle-ear transmission, i.e. $\delta = 0$ dB ($\kappa = 0.16$). The results for $V_C = 0.44$ cc correspond to using $\delta = +8$ dB, $\kappa = 0.40$. The values of the above parameters are summarized in Table 1 with estimates of the low-frequency equivalent compliance at the eardrum $V_d = V_e(P_{DC} = 0) - V_C = 1.24$ cc $- V_C$.

Table 1.

| δ (dB) | $20\log_{10}\kappa$ | κ | $V_C$ (cc) | $V_d$ (cc) |
|---|---|---|---|---|
| $-\infty$ | $-\infty$ | 0 | 0.76 | 0.48 |
| 0 | -16 | 0.16 | 0.66 | 0.58 |
| +5 | -11 | 0.28 | 0.57 | 0.67 |
| +8 | -8 | 0.40 | 0.44 | 0.80 |

For each estimate of $Y_d(f)$ in Figure 11, $|P_d(f)/P_{ec}(f)|$ was computed using Eq. 11.

The results in Fig. 11 are not exactly equivalent to processing each subject's data with the different $\delta$'s and then averaging the results across subjects (because Eq. 9 represents a nonlinear transformation). However, computations indicated that the results in Fig. 11 are, in fact, a very close approximation to those obtained when the individual data are (laboriously) processed with the different $\delta$'s.

The results in Fig. 11 illustrate that in some respects the influence of the different possible values of $V_C$ is substantial.

First, significant effects on the calculated values of $|P_d(f)/P_{ec}(f)|$ are seen to occur for the different $V_C$'s [see inset of Fig. 11 (a)]. The values suggested by Shaw for $|P_d(f)/P_{ec}(f)|$ increase with frequency to just over 10 dB at 3.8 kHz, and then fall for higher frequencies. In the (sensitive) region from 2 to 4 kHz, the values computed from $Y_d(f)$ associated with $V_C = 0.57$ cc (and $\delta = +5$ dB) provide a close fit to the Shaw-values. Furthermore, the values computed from the $Y_d(f)$'s associated with each of the other choices of $V_C$ (and $\delta$) are clearly in poorer agreement. In particular, $V_C = 7.6$ cc results in calculated values of $|P_d(f)/P_{ec}(f)|$ that are well below the Shaw-values; results such as these led us to question the assumption that $\delta = -\infty$ dB ($\kappa = 0$), and led to the choice of $\delta = +5$ dB.

Second, significant effects on $Y_d(f)$ are seen to occur for the different $V_C$'s [see Fig. 11 (a)]. At low frequencies, the estimates of $|Y_d(f)|$ and $V_C$ "trade" with each other as indicated by Eq. 8 (see also

Table 1). Above 1 kHz, $\measuredangle Y_d(f)$ is particularly affected. With increases in $V_C$, the resulting estimates of $\measuredangle Y_d(f)$ decrease. For example, for $V_C = 0.44$ cc, $\measuredangle Y_d(f)$ is always positive and is near +30 degrees between 2 and 4 kHz; in contrast, for $V_C = 0.76$ cc, $\measuredangle Y_d(f)$ is negative above 1.15 kHz and reaches -45 degrees near 4 kHz. In terms of the impedance components [Fig. 11 (b)]: At low frequencies, $R_d(f)$ shows a marked sensitivity to $V_C$; e.g. below 400 Hz, $R_d(f)$ for $V_C = 0.76$ cc is ~75% larger than $R_d(f)$ for $V_C = 0.57$ cc. Above 1 kHz, $X_d(f)$ is particularly sensitive to $V_C$, with the sign of $X_d(f)$ changing from negative to positive as $V_C$ decreases from 0.66 to 0.44 cc.

## III. C. Principal results on $Y_d(f)$

Using $\delta = +5$ dB and Eq. 10, $\kappa$'s were calculated for each subject. $V_C$ was then determined for each session using Eq. 9. The group-average $V_C$ was 0.57 cc (as noted in Sec. III. B); the individual-session $V_C$'s and the individual-subject $\kappa$'s are given in Table 2. Using Eq. 7 with $A_C = 0.48\ \text{cm}^2$, all measurements of $Y_e(f)$ were transformed to yield estimates of $Y_d(f)$.

The estimates of $Y_d(f)$ were then combined, as detailed below, (a) within sessions for each subject (intrasession), (b) across sessions for each subject (intersession or intrasubject), and (c) across subjects (intersubject).

For each session, all estimates of $Y_d(f)$ from sweeps in the "resting" condition (i.e. with no contralateral noise, with $P_{DC} = 0$, and with $60 \leqslant P_{REQ} \leqslant 75$ dB SPL) were averaged to obtain an intrasession mean $Y_d(f)$; an intrasession standard deviation, $\sigma[Y_d(f)]$, was also calculated. The number of sweeps entering into these intrasession results ranged from 3 to 7.

For each subject, the intrasession mean $Y_d(f)$'s were used to calculate an intrasubject mean $Y_d(f)$ and an intrasubject $\sigma[Y_d(f)]$. The number of sessions for each subject were 4 for S1, 3 for S2, 4 for S3, and 3 for S4. The sessions were separated by 2 to 10 weeks.

Table 2.

| Subject | $\Delta\|H\|$ (dB) | $\kappa$ | Session | $V_e(P_{DC}=0)$ (cc) | $V_e(P_{DC}=\pm40)$ (cc) | $V_C$ (cc) |
|---|---|---|---|---|---|---|
| S1 | -20.0 | 0.178 | 1 | 1.298 | 0.753 | 0.635 |
| | | | 2 | 1.327 | 0.837 | 0.731 |
| | | | 3 | 1.286 | 0.704 | 0.578 |
| | | | 4 | 1.172 | 0.622 | 0.503 |
| | | | Average | 1.271 | 0.729 | 0.612 |
| S2 | -14.5 | 0.335 | 1 | 1.037 | 0.670 | 0.485 |
| | | | 2 | 1.057 | 0.612 | 0.388 |
| | | | 3 | 1.232 | 0.756 | 0.516 |
| | | | Average | 1.109 | 0.679 | 0.463 |
| S3 | -15.0 | 0.316 | 1 | 1.225 | 0.730 | 0.501 |
| | | | 2 | 1.464 | 0.923 | 0.673 |
| | | | 3 | 1.039 | 0.620 | 0.426 |
| | | | 4 | 1.147 | 0.637 | 0.401 |
| | | | Average | 1.219 | 0.727 | 0.500 |
| S4 | -16.0 | 0.282 | 1 | 1.363 | 0.897 | 0.714 |
| | | | 2 | 1.414 | 0.892 | 0.687 |
| | | | 3 | 1.330 | 0.864 | 0.681 |
| | | | Average | 1.369 | 0.884 | 0.694 |
| $\overline{S}$ | -16.4 | 0.278 | | 1.242 | 0.755 | 0.567 |

An intersubject "group-average" $Y_d(f)$ was then calculated by averaging the intrasubject mean $Y_d(f)$'s for the four subjects; the intersubject $\sigma[Y_d(f)]$ was also calculated. In calculating these intersubject results, however, the results for subject S4 were not included up to 4 kHz because (see Sec. II. H) no probe-tube extension was usable with this subject. Somewhat arbitrarily, we have included S4's results on admittance magnitude for $f \leqslant 3$ kHz, and on admittance angle for $f \leqslant 2$ kHz.[16]

All of the above calculations were done with $Y_d(f)$ expressed as $|Y_d(f)|$ in dB re 1 mS (cgs) and $\measuredangle Y_d(f)$ in degrees. Thus, all $\sigma[Y_d(f)]$'s are given in dB and degrees.

The results are shown in Figs. 12 and 13 in terms of the magnitude and angle of the admittance at the eardrum, and in Figs. 14 and 15 in terms of the components of resistance, $R_d(f)$, and reactance, $X_d(f)$, of the impedance at the eardrum, $Z_d(f) = R_d(f) + jX_d(f) = 1/Y_d(f)$. The magnitude of the sound-pressure transformation of the earcanal, $|P_d(f)/P_{ec}(f)|$, calculated from the group-average $Y_d(f)$ (using Eq. 11) appears in Fig. 13 along with the values suggested by Shaw (1974a). Values for $|P_d(f)/P_{ec}(f)|$ calculated from each subject's (intersession) mean $Y_d(f)$ are given in Fig. 16. [Throughout these figures, results for subject S4 are shown up to the frequency limits noted above; specifically, results on $|Y_d(f)|$, $\sigma[|Y_d(f)|]$, $R_d(f)$, and $|P_d(f)/P_{ec}(f)|$ are shown up to 3 kHz, and results on $\measuredangle Y_d(f)$, $\sigma[\measuredangle Y_d(f)]$, and $X_d(f)$ are shown up to 2 kHz.]

To describe the results on the admittance at the eardrum (see Figs. 12 and 13), we consider separately the frequency regions below and above ~500 Hz.

Below 500 Hz, the admittance at the eardrum is compliance dominated; $|Y_d(f)|$ decreases approximately 6 dB per halving of frequency and $\measuredangle Y_d(f)$ approaches +90 degrees. For the group-average results, $\bar{S}$, a low-frequency equivalent compliance of 0.70 cc is indicated.[17] The individual-subject results for $|Y_d(f)|$ and $\measuredangle Y_d(f)$ are, respectively, within a fraction of a dB and a few degrees of those for $\bar{S}$.

Above 500 Hz, the admittance at the eardrum exhibits a pattern of undulations. $|Y_d(f)|$ exhibits a local maximum near 1.2 kHz, a local minimum near 2 kHz, and an increasing value as $f \rightarrow 4$ kHz. $\measuredangle Y_d(f)$ exhibits a local minimum near 1.6 kHz. In general, the local maxima and minima in $\measuredangle Y_d(f)$ occur at frequencies where points of inflection occur in $|Y_d(f)|$, and vice versa. While these undulations are clearly present for each subject, they indicate heavily damped "resonances" of the system for two reasons: First, the undulations in $Y_d(f)$ are relatively small; the largest "peak-to-peak" undulations occur for S1, about 7 dB for $|Y_d(f)|$ and 45 degrees for $\measuredangle Y_d(f)$ (above 1 kHz). Second, all the values of $\measuredangle Y_d(f)$ above 1 kHz are confined within ±45 degrees of zero indicating that the real part of $Y_d(f)$ exceeds the imaginary part $Y_d(f)$ in this frequency region. For the group-average results, $\bar{S}$ (see Fig. 13), the undulations in $Y_d(f)$ above 1 kHz are somewhat smoothed relative to those evident in the individual-subject data.

For $\bar{S}$ above 1 kHz, $|Y_d(f)|$ undulates about a value of +8 dB re 1 mS (2.5 mS) and $\measuredangle Y_d(f)$ is near zero. For the individual-subject results above 1 kHz, the values of $|Y_d(f)|$ are within ±3 dB of those for $\bar{S}$; the values of $\measuredangle Y_d(f)$ are within ±25 degrees of those for $\bar{S}$. Note also that while $\measuredangle Y_d(f)$ crosses zero and reaches minima of -15 to -35 degrees for S1, S3, and S4, the values of $\measuredangle Y_d(f)$ for S2 are all positive with the minimum value being +10 degrees.

For the results expressed in terms of the resistance, $R_d(f)$. and the reactance, $X_d(f)$, (see Figs. 14 and 15) the following features are prominent: Over the entire frequency range tested, $R_d(f)$ is within a relatively narrow range from about 250 to 700 ohms. Near 2 kHz, however, each subject exhibits a local increase in $R_d(f)$ of a few hundred ohms ($\sim$200 ohms for $\bar{S}$). Below about 300 Hz, $R_d(f)$ also increases. At low frequencies, $X_d(f)$ is large and negative reflecting the dominant complaince component identified in $Y_d(f)$. Above 1 kHz, $X_d(f)$ varies in size and sign from subject to subject, but the size of $X_d(f)$ is small compared to the size of $R_d(f)$ [and for $\bar{S}$, $X_d(f)$ is near zero, see Fig. 15].

The values of the sound-pressure transformation of the earcanal, $|P_d(f)/P_{ec}(f)|$, calculated from the group-average $Y_d(f)$ (see Fig. 13) are in close agreement with the values given by Shaw (1974a). [This agreement was already noted in the discussion of the choice of $\delta$ in Sec. III. B.] The values calculated from the individual-subject

$Y_d(f)$'s (see Fig. 16) exhibit deviations of ±3 dB from the Shaw-values; however, such deviations are typical of those reported to occur in measurements of $|P_d(f)/P_{ec}(f)|$ for individual subjects (Wiener and Ross, 1946; Djupesland and Zwislocki, 1973).

Some further results on the variability of our admittance measurements are given in Figs. 17 and 18. Fig. 17 shows that above 500 Hz, intersubject differences exceed intrasubject variability thus reinforcing an earlier remark that reproducible intersubject differences occur in the measurements of $Y_d(f)$ above 500 Hz. [Further comments on these results are given in Sec. IV. A. (3).] In Fig. 18 the variability in $Y_d(f)$ for individual subjects is compared across sessions (intersession) and within sessions (intrasession). Considering the four subjects, no simple ordering is apparent among these data, although all the $\sigma[Y_d(f)]$'s are relatively small. Notice, however, that S1 (the author) shows the smallest intrasession variability, typically a few tenths of a dB and a few degrees over the entire frequency range tested. We suspect at least two reasons may be responsible for the other subjects' exhibiting larger intrasession variability: (1) Although the subjects were instructed to remain stationary during the sweep measurements, S1 was probably more careful to do so. (2) Variations of the static pressure within the middle-ear space during the course of the sessions may have contributed to intrasession $\sigma[Y_d(f)]$. Specifically, we have found that changes in $Y_d(f)$ on the order of the size of the intrasession $\sigma[Y_d(f)]$'s in Fig. 18 occur in response to changes of the static pressure in the earcanal of ±2 cm $H_2O$ re ambient. [Our results on the effects of earcanal static pressure on $Y_d(f)$ are in preparation.] While all the sweeps entering into the $Y_d(f)$ results reported herein were made with ambient pressure in the earcanal, it is conceivable that variations of the middle-ear static pressure (of a few cm $H_2O$) occurred both from the ongoing absorption of oxygen by the tissues lining the middle-ear space, and also because the sessions included measurements with earcanal pressures up to ±40 cm $H_2O$ which (through displacements of the tympanic membrane) can be expected to alter middle-ear pressure. In Sec. II. H, it was indicated that the subjects were instructed to swallow prior to most sweeps in an attempt to insure near-zero middle-ear pressure. S1 was especially careful to swallow vigorously before every resting-condition sweep (with ambient earcanal pressure); the other subjects may not have been as vigorous,and as a result, may have had larger intrasession variations of middle-ear pressure. In two studies (Ithell, 1963; Lilly and Shepherd, 1964; see also, Lilly, 1973, pp. 360-361) it has been reported that over periods of 20 to 50 minutes steady and large changes sometimes occur in impedance measurements of the ear (particularly the resistance component). They hypothesized that these impedance changes over time may be due to changes in the sound absorption properties of the wall of the earcanal as a result of its accumulating moisture. Our sessions lasted 20 to 75 minutes; no such changes are apparent in our measurements, and given our comments above, the "true" intrasession variability in $Y_d(f)$ may be no larger than that obtained for S1.

## III. D. Influence of the inflatable cuff on $Y_d(f)$

For the calibration of the EPA in the precision cavities (see Sec. II. B) the silicone-fluid-filled inflatable-cuff was removed from the EPA-earpiece. In using the $Y_s(f)$ determined with the cuff absent, we have implicitly assumed that the cuff has no influence on the admittance measured at the EPA-tip; i.e., we have assumed that the cuff acts like a rigid boundary with zero admittance. To test this assumption, an experiment was performed on subject S1 using a hard (rubber-like) eartip fitted on the EPA-earpiece in place of the inflatable cuff. The hard eartip is assumed to provide a good approximation to a rigid boundary.[18]

A probe-tube extension of 3.0 mm was used in the experiment with the hard eartip. The measurements from this experiment were processed exactly like those from the sessions with the inflatable cuff; i.e. the measurements of $V_{pm}(f)/V_i(f)$ were transformed to estimates of $Y_e(f)$, $V_C$ was determined, and then the $Y_e(f)$'s were transformed to estimates of $Y_d(f)$.

In Fig. 19 the average $Y_d(f)$ from the experiment with the hard eartip is shown along with the average intersession $Y_d(f)$ for S1 obtained from the sessions using the inflatable cuff. The differences between the two $Y_d(f)$'s are small; thus, the inflatable-cuff appears to have little influence on the measurements of $Y_d(f)$.

## IV. Discussion

### IV. A. Comparisons to other data

#### (1) Laboratory data

In this section our measurements of $Y_d(f)$ are compared to certain other "laboratory" studies in which results were obtained over relatively wide frequency ranges, and particularly above 1 kHz. [Further review is given in the studies cited below and in Sec. IV. A. (2).]

Møller (1960) reported measurements of $Y_d(f)$ from 200 Hz to 2 kHz using methods that were similar to ours, but with two differences: First, the volume $V_C$ was measured by filling the residual earcanal space with alcohol (from a calibrated syringe). Second, in transforming the admittance values in the reference plane of measurement, "$Y_e(f)$", to estimates of $Y_d(f)$, Møller treated the space $V_C$ as a lumped acoustic compliance, i.e. Eq. 8 was used, whereas we have treated this space as a section of a lossless acoustic transmission line, i.e. Eq. 7 was used. With increasing frequency, increasing differences accrue between the $Y_d(f)$'s computed by these two procedures; e.g. at 2 kHz, the estimates of $Y_d(f)$ from Eqs. 7 and 8 for a typical $Y_e(f)$ shown by Møller (1960, Fig. 4, subject BL) differ by 2.3 dB in magnitude and 21 degrees in angle.

Zwislocki (1970) reported measurements of the impedance at the eardrum, $Z_d(f) = R_d(f) + jX_d(f)$; $R_d(f)$ was measured from 200 Hz to 4 kHz and $X_d(f)$ was measured from 200 Hz to 7 kHz. The measurements were obtained using a Zwislocki acoustic bridge (Grason-Stadler E8872A) that (a) was specially calibrated to extend the upper frequency limit of operation beyond the 1 to 1.5 kHz limit normally associated with this instrument, and (b) was modified to permit positive eardrum reactances to be measured. The volume $V_C$ was determined by liquid-filling. The adjustable arm of the bridge incorporates a tube section of cross-sectional area approximating that of the earcanal (value not given) and of length which is adjusted to obtain an enclosed volume equal to $V_C$.

Blauert and Platte (1974) reported estimates of $Y_d(f)$ that were suggested to be reliable from about 1.3 to 8 kHz. These estimates were obtained using an "impulse-response" method (as detailed in Blauert et al, 1974) in which free-field acoustic impulses were presented to the ear and time-domain measurements of the sound-pressure waveform were made (using a probe-tube microphone) first at a location near the eardrum, $p_d(t)$, and then at the entrance to the earcanal, $p_{ec}(t)$. These responses were Fast-Fourier-Transformed and then divided to obtain the magnitude and angle of the earcanal sound-pressure transformation, $P_d(f)/P_{ec}(f)$. Estimates of $Y_d(f)$ followed using Eq. 11 with $L_{EC}$ obtained as the probe-tube microphone was retracted from the eardrum position to the earcanal entrance. Although the estimates of $Y_d(f)$ were calculated from 0 to 16 kHz, Blauert and Platte indicated that (a) below 1.3 kHz, there is poor precision in determining $Y_d(f)$ because $P_d(f)/P_{ec}(f) \rightarrow 1$, and (b) above 8 kHz, there was a poor signal-to-ratio in the measurements.

Comparing the results of these three studies to our average results we observe: (1) At low frequencies, where the magnitude of the reactance $X_d(f)$ exceeds the resistance $R_d(f)$, there is close agreement for the values of $X_d(f)$ [see Fig. 20(a)]. (2) Above 1 kHz, the values of $X_d(f)$ are generally near zero [$|X_d(f)| < 200\ \Omega$]. (3) Above 1 kHz, significant differences exist in the pattern vs frequency and the values of $R_d(f)$. The local increase in $R_d(f)$ near 2 kHz that is prominent in our data is not evident in the results from the other studies. Additionally, above ~1.5 kHz, our values for $R_d(f)$ exceed those reported in the other studies; e.g. near 2 kHz, our $R_d(f) \approx 500\ \Omega$ whereas the $R_d(f)$'s in the other studies range from 200 to 250 Ω, and near 4 kHz, our $R_d(f)$ 300 Ω whereas Blauert and Platte show 140 Ω and Zwislocki shows 45 Ω.

The earcanal sound-pressure transformation, $|P_d(f)/P_{ec}(f)|$, can be calculated from the $Y_d(f)$ from each study [see Fig. 20(b)]. Zwislocki's results are inconsistent with the ∼10 dB peak in $|P_d(f)/P_{ec}(f)|$ near 3.8 kHz. Zwislocki (1970, Sec. 4.3) recognized this discrepancy; he concluded that his $R_d(f)$ measurements were "faulty" in this frequency region, and furthermore, that in order to reduce this discrepancy in $|P_d(f)/P_{ec}(f)|$, larger values of $R_d(f)$ were needed, assuming $X_d(f)$ is near zero.

Fig. 20(b) also shows a measurement of $|Y_d(f)|$ obtained on a "fresh" human cadaver by Fischler et al (1966). Using an instrument incorporating a capacitive-probe vibration indicator and a probe-tube microphone, the displacement magnitude of the eardrum in the vicinity of the umbo, "$|X_{umbo}(f)|$", and the sound-pressure magnitude at the eardrum, $|P_d(f)|$, were measured from 40 Hz to 10 kHz. The measurements were presented in terms of the mechanical impedance at the eardrum, $2\pi f|X_{umbo}(f)/P_d(f)|$; we converted these values to estimates of $|Y_d(f)|$ by multiplying them by $A_C$ = earcanal area = 0.48 $cm^2$ and then taking the reciprocal. The very low $|Y_d(f)|$ at low frequencies for the Fischler et al results suggests that postmortem changes may have occurred in their cadaver specimen (Zwislocki and Feldman, 1963); thus, interpretation of their results with respect to living humans is difficult.

### (2) Values suggested by Shaw (1975b)

Shaw (1975b) recently suggested representative values for the impedance at the eardrum, $R_d(f)$ and $X_d(f)$; these values (see Fig. 21) were based on different considerations for frequencies below and above ~1.5 kHz.

Below 1.5 kHz, Shaw fitted average or median results reported in a number of studies. For most of these results, the volume of the residual earcanal space, $V_C$, was determined by liquid-filling. At low frequencies, our results are in close agreement with those given by Shaw. This agreement cannot, however, be construed as strong support for the particular choice of $\delta = +5$ dB used in our "indirect" method for determining $V_C$ (see Sec. III. B). Specifically, had we used $\delta = +8$ or 0 dB, our results [see Fig. 11(b) with $V_C = 0.44$ and 0.66 cc, respectively] would be within the approximate range of the results fitted by Shaw. The "limiting-case" $\delta = -\infty$ dB ($\kappa = 0$) leads, however, to estimates of $R_d(f)$ that are outside this range. For example, near 250 Hz, the results for $R_d(f)$ fitted by Shaw range from about 250 to 550 Ω; the results in Fig. 11(b) show $R_d(f) = 825$ Ω for $V_C = 0.76$ cc ($\kappa = 0$). These comparisons provide further evidence (see Sec. III. B) against the assumption that $\kappa = 0$.

Above 1.5 kHz, Shaw considered reported measurements of other acoustical properties of the external ear, and in particular, measurements of the "standing-wave-ratio" (SWR) in the earcanal. He recognized that if $R_d(f) >> |X_d(f)|$, then SWR $\simeq R_d(f)/(\rho c/A_C)$ where $\rho c/A_C$ is the characteristic acoustic impedance associated with the earcanal. For the first (quarter-wave) resonance of the earcanal near 3.8 kHz, SWR $= |P_d(f)/P_{ec}(f)| \simeq 10.5$ dB which leads to $R_d/(\rho c/A_C) \simeq 3.3$. Additionally, Shaw considered the first resonance of the external ear, including the earcanal and the concha; this resonance occurs near 2.7 kHz and has a SWR $\simeq 14$ dB which leads to $R_d/(\rho c/A_C) \simeq 5$. Using $A_C = 0.44$ cm$^2$ as the average earcanal area for adult males and females, and therefore that $\rho c/A_C \simeq 92$ Ω (for our male subjects $A_C = 0.48$ cm$^2$ yields $\rho c/A_C \simeq 85$ Ω), Shaw suggested that $R_d(f)$ should be in the vicinity of 400 Ω at 2.7 kHz and 300 Ω at 3.8 kHz. These values exceeded those indicated from available measurements of $R_d(f)$ [c.f. Sec. IV. A. (1)]; Shaw concluded the available measurements were "clearly unreliable". The values are, however, close to those indicated by our measurements, although the "bulge" of $R_d(f)$ in our measurements (near 2 kHz) extends to frequencies below that suggested by Shaw.

### (3) Clinical data

In this section, our measurements of $R_d(f)$ and $X_d(f)$ are compared to certain "clinical" studies in which results were obtained on normal ears using admittance measuring devices that are commercially available. We consider separately measurements obtained (a) using the Zwislocki acoustic bridge, and (b) using "tympanometry". [Further review is given in the studies cited below and in Jerger, 1975; Feldman and Wilber, 1976; Northern, 1976.]

In using the Zwislocki acoustic bridge, it has been reported (c.f., Burke, Nilges, and Henry, 1970; Zwislocki and Feldman, 1970) that a major source of error results from inaccuracies (a) in the measurement of the residual earcanal space $V_C$ by liquid-filling, and (b) in attempting to reinsert the speculum to the same position in the ear [which existed for (a)] for the subsequent measurements of the impedance of the ear. Burke, Nilges, and Henry (1970) report that the "total" error [(a) + (b)] is "easily" ±0.1 cc for a skillful operator. We compare (in Fig. 22) our measurements of $R_d(f)$ and $X_d(f)$ only to those reported by Burke, Nilges, and Henry (1970) and Zwislocki and Feldman (1970); we presume that these particular operators were "skillful" in their experimental technique.

The mean (median) values for $R_d(f)$ and $X_d(f)$ from these two studies and our average results are in general agreement, but two differences are indicated: (1) At low frequencies, our $\bar{S}$ values for

$X_d(f)$ are smaller (in magnitude) than the mean (median) values from the two Zwislocki-bridge studies; e.g. at 125 and 250 Hz, the difference is about 17%. This difference occurs partly because our subjects were all males whereas the clinical studies included males and females (numbers not given); at low frequencies, Zwislocki (1970, Fig. 24) indicates that the median $|X_d(f)|$ for males is about 24% smaller than that for females (see also Møller, 1960, Fig. 3). Additionally, the size of the intersubject variability shown for the clinical data (which are for considerably larger subject populations than our study) suggests that the differences in the means may not be significant. The relatively small intersubject differences exhibited by our results may reflect a chance occurrence (only 4 subjects were tested); however, it partially derives from our procedure for determining $V_C$. In particular, our assumption (Eq. 10) that $|P_{DC}| = 40$ cm $H_2O$ causes proportional effects on the low-frequency admittance-change and transmission-change for each subject resulted in smaller intersubject differences occurring for $X_d(f)$ [and for $|Y_d(f)|$] at low frequencies than would have occurred had we used a single value for $\kappa$. (2) At 1.5 kHz, Burke, Nilges, and Henry indicate a mean reactance (-207 Ω) which is equal to the lower limit of our results, and furthermore, they indicate an intersubject standard deviation (73 Ω) which is small compared to the range of our results. However, their results for $X_d(f)$ at 1.5 kHz are erroneous because they failed to account for the positive reactance associated with the adjustable resistance element of the bridge. This reactance ($X_R$) increases with the value of the resistance (R) and with frequency; Zwislocki (1970, Figs. 12 and 13) indicates that for $R \simeq 150$ to 500 Ω, $X_R \simeq +90$ to +250 Ω at 1.5 kHz, $X_R \simeq +50$ to +70 Ω at 750 Hz, and $X_R$ is near zero for f $\lesssim$ 500 Hz. $X_R$ acts in "series" with the negative reactance of the adjustable "eardrum" compliance element of the bridge; hence; the values for $X_d(f)$ at 1.5 kHz reported by Burke, Nilges, and Henry (and to a lesser extent, the values reported at lower f's from both Zwislocki-bridge-studies) are erroneously biased toward nagative reactance values which are too large. [Results from other studies using the Zwislocki bridge (Tillman et al, 1964; Bicknell and Morgan, 1968; Djupesland and Kvernvold, 1975, and others) are similarly in error.]

Tympanometry (as currently used) involves changing the static pressure in the earcanal ($P_{DC}$) and simultaneously measuring the variations of the admittance of the ear ($Y_e$) at one or two (low) frequencies. One underlying assumption of tympanometry is that the admittance at the eardrum ($Y_d$) is a maximum when $P_{DC}$ equals the pressure in the middle ear ($P_{ME}$), and that for $P_{DC}$ increasingly different from $P_{ME}$, $Y_d$ decreases.

Several instruments are commercially available for making such measurements. Most of the instruments operate ata single low frequency, ~220 Hz, and measure only admittance magnitude, i.e. $|Y_e|$ vs $P_{DC}$; one instrument (Grason-Stadler 1720 Otoadmittance Meter) operates at two frequencies, 220 and 660 Hz, and measures the conductance $G_e$ and susceptance $B_e$ (vs $P_{DC}$). Various qualitative and quantitative descriptions of tympanograms obtained with these instruments have been demonstrated to be clinically useful for the diagnosis of pathologies affecting the middle ear (c.f. Secs. II, V, and VI of Northern, 1976).

While the measurement of an admittance quantity (or quantities) vs $P_{DC}$ is universal to tympanometry, standards have not been developed (a) for some experimental variables involved in measuring tympanograms, e.g. the direction and speed of changing $P_{DC}$, and (b) for calculating $Y_d$ from tympanograms, e.g. which points are selected as representing (1) $Y_d + Y_C$ and (2) $Y_C$ alone, where $Y_C$ is the admittance of the residual earcanal space. The influences of some experimental parameters and computational procedures (on the estimation of $Y_d$) have been studied (Porter and Winston, 1973a, 1973b; Creten and Van Camp, 1974; Margolis and Smith, 1976) and comparisons between $Y_d$-estimates (at 220 and/or 250 Hz) obtained using tympanometry and using the Zwislocki acoustic bridge on the same subjects have been reported (Burke, Herer, and McPherson, 1970; Feldman et al, 1971). From these studies some questions have arisen concerning the interpretation of $Y_d$-estimates from tympanometry. In particular, Creten and Van Camp (1974) report that when the speed of changing $P_{DC}$ (i.e. $dP_{DC}/dt$) is reduced from the typical clinically-used rate of ~3 cm $H_2O$/sec to 0.1 cm $H_2O$/sec, the peak values of tympanograms (measured using a GS-1720 Otoadmittance Meter) "decreased sometimes as much as 50%". Thus, we will not compare particular values of $Y_d$ obtained using tympanometry to our results; however, we will comment on certain differences between our procedure for transforming $Y_e(f)$ to $Y_d(f)$ and that used in tympanometry.

First, in using tympanometry it is routinely assumed that large $P_{DC}$ reduces $Y_d(f)$ to zero; i.e. in terms of the notation in Sec. III. B, it is assumed $\kappa = 0$. As noted in Sec. III. B, we have not assumed $\kappa = 0$; on consideration of measurements of middle-ear transmission-change vs $P_{DC}$ at low frequencies and representative values of the ear-canal sound-pressure transformation between ~2 and 4 kHz, $\kappa$ was determined to be 0.28 (for $\bar{S}$). Furthermore, in terms of low-frequency equivalent compliances, our average results indicate that "eardrum" compliance contributes 0.19 cc to the compliance measured for $|P_{DC}| = 40$ cm $H_2O$ [i.e. from Table 2 in Sec. III. B, $V_e(|P_{DC}|=40) - V_C = 0.76 - 0.57 = 0.19$ cc]. Because our results have been obtained on four subjects, it is unclear whether the particular values of $\kappa$ (and $\delta$) that we determined are applicable to a larger population. Nevertheless, as was noted in Sec. IV. A. (2), had we assumed $\kappa = 0$ our results for $R_d(f)$ at low frequencies would be outside the range of results reviewed by Shaw (1975b).

Second, when tympanometry is performed at 220 and 660 Hz using the GS-1720 Otoadmittance Meter, the admittance of the residual earcanal space $Y_C(f)$ is estimated for each frequency from the values for $G_e$ and $B_e$ measured with large $P_{DC}$. We have transformed $Y_e(f)$ to $Y_d(f)$ for all frequencies using a single earcanal volume $V_C$ (and earcanal area $A_C$) in conjunction with a lossless transmission-line transformation (Eq. 7). [A similar transformation occurs with the Zwislocki acoustic bridge.] At low frequencies, our transformation can be approximated as (Eq. 8): $Y_d(f) = Y_e(f) - Y_C(f)$, where $Y_C(f) = G_C(f) + jB_C(f)$, and $G_C(f) = 0$ and $B_C(f) = j2\pi fV_C/\rho c^2$. Thus, in our processing $G_C(220\text{ Hz}) = G_C(660\text{ Hz}) \approx 0$ and $B_C(660\text{ Hz})/B_C(220\text{ Hz}) \approx 3$, whereas no relation between $Y_C(220\text{ Hz})$

and $Y_C$(660 Hz) is imposed in clinical tympanometry.

We have indicated certain differences between our procedure and the procedure(s) used with tympanometry for transforming measurements of $Y_e(f)$ to estimates of $Y_d(f)$. Both procedures involve assumptions which influence the resulting estimates of $Y_d(f)$; we suggest that the assumptions warrant further consideration. In addition, if one considers using tympanometry at frequencies above 660 Hz, we suggest that the assumption that $Y_d(f) = 0$ for large $P_{DC}$ is clearly suspect for two reasons: (1) Our measurements of $Y_e(f)$ [see Sec. III. A and Fig. 10] indicate that above ~1.5 kHz, large $P_{DC}$ does not reduce $|Y_e(f)|$, but instead, $|Y_e(f;P_{DC}=+40)| > |Y_e(f;P_{DC}=-40)| \geqslant |Y_e(f;P_{DC}=0)|$. (2) Our measurements of middle-ear transmission-change vs $P_{DC}$ indicate that the transmission reductions which occur for $2 \leqslant f \leqslant 4$ kHz with $|P_{DC}| = 40$ cm $H_2O$ are relatively small (< 5 dB).

## IV. B. Comparison to the Zwislocki (1962) middle-ear model

Zwislocki (1962) developed a circuit-model analog for the human middle-ear system. The structure of the model was based on considerations of the functional anatomy of the middle-ear system; element values for the model were primarily determined from acoustic impedance measurements and were chosen to be representative of an "average" ear.

Fig. 23 shows that the values of $Y_d(f)$ calculated from the Zwislocki model and our average results for $Y_d(f)$ exhibit generally similar shapes vs frequency. However, two differences between the model calculations and our measurements are evident: (1) The $|Y_d(f)|$ at low frequencies indicated by our measurements is 2 to 3 dB below that calculated from the model. (2) The undulations of $Y_d(f)$ above 1 kHz in our measurements occur a few tenths of an octave below the corresponding undulations of $Y_d(f)$ for the model. Neither of these discrepancies invalidates the structure of the Zwislocki model; it is likely that modest changes in some element values could bring $Y_d(f)$ for the model into closer agreement with our results.

Near 2.3 kHz, the Zwislocki model exhibits pronounced effects in $Z_d(f)$ due to a resonance within the middle-ear cavities. These effects are manifest as (a) a local increase in $R_d(f)$, and (b) an increase in $X_d(f)$ below 2.3 kHz and a decrease in $X_d(f)$ above 2.3 kHz. In selecting element values for the middle-ear-cavity analog, Zwislocki made use of Onchi's (1960) impedance measurements reported for a single cadaver specimen. Because reliable measurements of $Z_d(f)$ in living humans have been restricted to frequencies below ~1.5 kHz [see Secs. IV. A. (1) and (2)], the variations in $Z_d(f)$ near 2.3 kHz predicted from the Zwislocki model have not been demonstrated; our measurements indicate a frequency dependence of $Z_d(f)$ that is similar to that predicted from the model. Finally, we reiterate that Shaw's (1975b) analysis of acoustical measurements on the external ear also leads to a predicted increase in $R_d(f)$ in this frequency region [see Sec. IV. A. (2)].

## V. Summary

An acoustic system was designed for measuring the acoustic admittance of the ear over a wide frequency range. The system was calibrated from 62 Hz to 4 kHz in terms of a Norton equivalent model; in particular, the source admittance $Y_s(f)$ was determined for each frequency.

The admittance at the eardrum, $Y_d(f)$, was estimated from measurements of the admittance in the earcanal, $Y_e(f)$, using a lossless transmission-line transformation requiring a single parameter $V_C$ to be determined for each session, where $V_C$ = the residual earcanal volume between the tip of the measuring device and the eardrum. $V_C$ was determined making use of (a) the low-frequency equivalent compliances indicated from $Y_e(f)$ with earcanal static pressure $P_{DC} = 0$, +40, and -40 cm $H_2O$, (b) the low-frequency (middle-ear) transmission-change resulting from $P_{DC} = \pm 40$ cm $H_2O$, and (c) reported measurements of the earcanal sound-pressure transformation, $|P_d(f)/P_{ec}(f)|$, between about 2 and 4 kHz. We did not assume that $P_{DC} = \pm 40$ cm $H_2O$ reduces $Y_d(f)$ to zero; our analysis indicated that $P_{DC} = \pm 40$ cm $H_2O$ reduces $|Y_d(f)|$ at low frequencies by about 11 dB (on the average).

Our results indicated that below 500 Hz, $Y_d(f)$ is compliance dominated, and that from 1 to 4 kHz, the resistance component of the impedance at the eardrum exceeds the magnitude of the reactance component. The results at low frequencies are in agreement with those reported by others. Our values for the resistance above 1 kHz exceed those measured by others; however, our values generally agree with values predicted from Shaw's (1975b) analysis of acoustical measurements on the external ear.

Our results are reconcilable with "clinical" measurements obtained using the Zwislocki acoustic bridge. Concerning tympanometry, we point out that two assumptions usually made with tympanometry, namely that $Y_d(f) = 0$ for large $P_{DC}$ and that the admittance of the residual earcanal space is not related for different frequencies, warrant further consideration.

## FOOTNOTES

1. We have borrowed extensively from the ideas of other designers of electroacoustic admittance-measuring devices, including Zwislocki (1957), Møller (1960), and Pinto and Dallos (1968).

2. The following notational conventions are used throughout. "Terminal variables" such as voltages, currents, sound pressures, and volume velocities are denoted by the complex root-mean-square amplitudes of steady-state sinusoidal waveforms; i.e., sinusoidal functions of the form $\sqrt{2}\,|V|\sin(2\pi ft + \measuredangle V)$ are represented by the complex amplitude $V(f)$, or simply $V$, where $V = |V|\exp(j\measuredangle V)$. "System variables" such as admittances, impedances, and other transfer functions denote ratios of two terminal variables and are also represented by complex amplitudes.

3. The sensitivity and offset circuit consists simply of a resistance bridge with the transducer in one bridge arm. The input to the bridge is an adjustable DC voltage, adjustment of which provides sensitivity control. Offset adjustment is accomplished by varying the resistance of the bridge arm opposite to the transducer. The output of the bridge is fed directly (with no amplification) to the voltmeter.

4. Eq. 4 follows from recognizing that:

$$v_m(t) = \sqrt{2}|V_m|\cos(2\pi ft)$$

$$v_q(t) = \sqrt{2}|V_q|\cos(2\pi ft + \measuredangle 90°) = -\sqrt{2}|V_q|\sin(2\pi ft)$$

and by denoting the signal from the output amplifier as $x(t)$, i.e. $X = A_oV_o$, and:

$$x(t) = \sqrt{2}|A_oV_o|\cos(2\pi ft + \measuredangle A_o + \measuredangle V_o)$$

and making use of the trignometric identities:

$$2\cos(a)\cos(b) = \cos(a+b) + \cos(a-b)$$

$$2\sin(a)\cos(b) = \sin(a+b) + \sin(a-b).$$

Specifically, the multiplier outputs are:

$$x(t)v_m(t) = |A_oV_oV_m|\cos(\measuredangle A_o + \measuredangle V_o) + |A_oV_oV_m|\cos(4\pi ft + \measuredangle A_o + \measuredangle V_o$$

$$x(t)v_q(t) = |A_oV_oV_q|\sin(\measuredangle A_o + \measuredangle V_o) - |A_oV_oV_q|\sin(4\pi ft + \measuredangle A_o + \measuredangle V_o$$

where the first terms are the DC components given by Eq. 4 and the second terms are the second-harmonic components which are removed by the low-pass filters.

5. This measurement scheme is analogous to an optimal synchronous detector for sinusoidal signals of unknown amplitude and phase in the presence of additive white noise (c.f., Secs. 7.2 and 7.3 of Wozencraft and Jacobs, 1965).

6. This averaging was done to reduce the effects of (relatively) rapid and irregular fluctuations of the microphone output which occur during the acoustic reflex.

7. From now on the frequency argument is explicitly represented.

8. $\xi$ is given by $\mu^{1/2} + (\gamma-1)[\nu/\gamma]^{1/2}$, where $\mu = 0.156$ cm$^2$/sec = kinematic coefficient of viscosity, $\nu = 0.27$ cm$^2$/sec = coefficient of temperature exchange, and $\gamma = 1.4$ = ratio of the specific heat at constant pressure to the specific heat at constant volume.

9. For example, these differences would be reduced if both $\alpha$ and $c'$ were increased re that given in Eq. 4. A decrease in the loss term $\xi$ would cause $c'$ to increase but $\alpha$ would decrease. However, a recent "exact" analysis of the acoustics of lossy tubes (Mook, 1976) indicates that the "true" $\alpha$ is underestimated by Eq. 4.

10. The order of these three sweeps varied from session to session and also between the initial and concluding sweeps of a given session.

11. The static pressures of +40 and −40 cm $H_2O$ cause perturbations to the middle-ear system resulting in a new "small-signal" operating point. A pressure change having an effective magnitude of 40 cm $H_2O$ would have a sound-pressure level of 166 dB SPL.

12. We have used the average of the compliances measured with $P_{DC}$ = +40 and -40 cm $H_2O$, although at least three factors could cause $V_e(P_{DC} = +40)$ to differ from $V_e(P_{DC} = -40)$: First, the acoustic compliance at the eardrum may differ for $P_{DC}$ = +40 and -40 cm $H_2O$. The use of a single $\kappa$ representing the effect of $|P_{DC}| = 40$ is an approximation we make because we are unaware of any measurements of (differential) effects of these two pressures on $V_d$. Second, with changes in $P_{DC}$, "static" displacement of the eardrum occurs; positive pressure displaces the eardrum inward thereby increasing $V_C$ re that occurring at ambient pressure, and the opposite occurs for negative pressure. However, measurements of eardrum volumetric displacement on 4 normal ears for $|P_{DC}|$ up to 30 cm $H_2O$ by Ingelstedt et al (1967, Fig. 9 shows results on 1 normal ear and Fig. 16 shows results on 3 pilots with normal ears) and similar measurements on 101 normal ears for $|P_{DC}|$ up to 15 cm $H_2O$ by Elner et al (1971) indicate that the eardrum displacements for positive and negative pressure are symmetric to within 0.01 cc. Therefore, the pressure effects on $V_C$ are compensated for by averaging $V_e(P_{DC} = +40)$ and $V_e(P_{DC} = -40)$. Third, the compliance C of a fixed volume V is influenced by the total static pressure of the air in the volume. Specifically, $C = V/\rho c^2 = V/\gamma P_T$ where $\gamma = 1.4$ and $P_T = P_0 + P_{DC}$ with $P_0$ = ambient atmospheric pressure ≈ 1030 cm $H_2O$. Thus, when $P_{DC}$ changes from zero to ±40 cm $H_2O$, $P_T$ changes about ±4%, and C changes ∓4%. For all calculations we have used values for $\rho$ and c at ambient pressure; however, the resulting errors that are introduced are cancelled by averaging the compliances measured for $P_{DC}$ = +40 and -40 cm $H_2O$.

13. With some commercial instruments the real and imaginary components of $Y_e$ are measured, $Y_e = G_e + jB_e$. In using these instruments, $Y_d = G_d + jB_d$ is assumed to be zero for large $P_{DC}$, and therefore, the admittance of the residual earcanal space is assumed equal to $Y_e$(large $P_{DC}$) which has both real and imaginary components.

14. The psychophysical methods used to obtain these measurements are described elsewhere (see Fig. 13 of Part 2).

15. Further details on the averaging that was done are given in Sec. III. C.

16. Analogous to the influence of different $V_C$'s on estimates of $Y_d(f)$ demonstrated in Fig. 11, from 2 to 4 kHz errors (or variations) in $Y_e(f)$ cause relatively large variations in $\measuredangle Y_d(f)$ and smaller variations in $|Y_d(f)|$.

17. This value differs slightly from the estimate of 0.67 cc given in Table 1 of Sec. III. B because the values for "$V_d$" in that table were based on the approximation that $V_d = V_e(P_{DC} = 0) - V_C$, which is exact only if $\measuredangle Y_d = \measuredangle Y_e = +90$ degrees.

18. The hard eartip was not used regularly because (a) based on preliminary tests it was judged more uncomfortable than the inflatable cuff, and therefore was less suitable for the relatively long sessions we had planned, and (b) it was difficult to insert the

hard-eartip/earpiece assembly into the earcanal to obtain a hermetic seal (essentially by a "force-fit") and not poke the earcanal wall with the probe-tube extension.

# FIGURE CAPTIONS

Fig. 1. Schematic diagram of the electroacoustic-probe-assembly (EPA) system; Norton equivalent model shown in the dotted insert of the lower right. See text for details; drawing not to scale.

Fig. 2. Block diagram of apparatus used for measuring and storing frequency responses.

Fig. 3. Timing diagram for the events within each tone-pulse cycle. Seventy-one tone-pulses per sweep.

Fig. 4. Sound-pressure level developed at a calibrated, reference microphone joined (hermetically) to an EPA. [0.7 cc enclosed between the EPA tip and the reference microphone.] ■: "Available" sound pressure, in dB SPL (0 dB SPL = 0.0002 $\text{dynes/cm}^2$), for maximum input voltage applied to the EPA, i.e. $|A_i| = 0$ dB. ▲: Sound-pressure level developed when 75 dB SPL is "requested", i.e. when $|A_i(f)|$, in dB, is set to the values indicated by ■ minus 75 (when possible). The dashed line shows a segment of the "available" sound-pressure level response when the high-pass compensation ($f_{-3\ \mathrm{dB}} \simeq 1300$ Hz) in the equalized earphone driver (see Fig. 1) is removed. At lower frequencies, the difference between this response and ■ increases 6 dB per halving of frequency.

Fig. 5. EPA probe-microphone output, $|V_{pm}(f)|$, for various load conditions. Input voltage to the EPA, $V_1(f)$, was the same for all conditions, and was specified as $P_{REQ} = 72$ dB SPL. Results shown for the measurements in the three precision cavities, as indicated, along with results from two additional test conditions: For "FREE RADIATION", the EPA-tip was left open and pointed away from all nearby surfaces. For "PLUGGED PROBE-TIP", a tapered wire was pressed into the tip of the microphone probe-tube and the EPA-tip was reinserted into the 1.0 cc cavity.

Fig. 6. Estimates of the Norton source admittance of the EPA, $Y_s(f)$. Admittance magnitude below, $|Y_s(f)|$ in dB re 1 cgs mS = $20\log_{10}[|Y_s(f)|/1 \text{ mS}]$, where 1 cgs acoustic mho = $1 \text{ cm}^5\text{dyne}^{-1}\text{sec}^{-1}$, and mS = millisiemen = mmho; admittance angle above, $\measuredangle Y_s(f)$ in degrees, and shown only for $f \leq 4$ kHz. The theoretical admittance magnitude of a 1 cc cavity, of 1 cm length, is also shown.

Fig. 7. Errors or "deviations" from a single Norton equivalent model for the EPA implied by the results for the three precision cavities. See text for details. Magnitude errors, in dB, above; angle errors in degrees, below. The solid lines indicate the errors between the calculated admittance implied by the measurements in the 0.5 c (▲) and 2.0 cc (▼) cavities and their respective theoretical admittance values. These results are shown only for $f \leq 5.6$ kHz;

at higher frequencies the errors increase rapidly. The dashed lines indicate the errors which result in measuring the admittances of the 0.5 cc (■) and 2.0 cc (◆) cavities if $Y_s(f)$ is assumed to be zero. At frequencies above which the dashed results are plotted, the dashed results indicate errors of the approximate size given by the solid lines.

Fig. 8. Narrow-tube admittance test. Comparison between the measured and theoretical admittance of two tubes, $Y_{2tu}(f)$; each tube of inner diameter D = 0.211 cm and length L = 43.1 cm. See text for details. Admittance magnitude, $|Y_{2tu}(f)|$ in dB re 1 mS, above; admittance angle, $\measuredangle Y_{2tu}(f)$ in degrees, below.

Fig. 9. Sound-pressure level in the earcanal at the tip of the EPA, $|P_e(f)|$, for different values of the requested sound-pressure level, $P_{REQ}$. $|P_e(f)|$ and $P_{REQ}$ in dB SPL. "NO SIGNAL" indicates the measured $|P_e(f)|$ when the input voltage to the EPA was removed, i.e. $V_i(f) = 0$. The results for $P_{REQ} = 75$, indicated —○—, are also shown shifted vertically in successive 10 dB steps and plotted as dots. Subject S1; EPA-microphone probe-tube extension 3.5 mm.

Fig. 10. Admittance of the ear in the plane of the EPA-tip, $Y_e(f)$. Typical results for two subjects: (a) S1, using an EPA with a microphone probe-tube extension of 3.5 mm, and (b) S3, with an

extension of 2.0 mm. Solid lines (marked with symbols) show $Y_e(f)$ for the three sweeps at the start of the session with earcanal static pressure $P_{DC}$ = 0, +40, and -40 cm $H_2O$; dashed lines (nearly superimposed on the solid lines) show $Y_e(f)$ for the three sweeps at the end of the session, again with $P_{DC}$ = 0, +40, and -40 cm $H_2O$. Session durations were 50 minutes for (a) and 60 minutes for (b). Admittance angle, $\measuredangle Y_e(f)$ in degrees above; admittance magnitude, $|Y_e(f)|$ in dB re 1 mS (cgs), below. Also shown is the theoretical admittance of a 1.0 cc cavity, of 1.0 cm length, $Y_{1\ cc}(f)$ plotted as dots, and the magnitude of the Norton source admittance, $|Y_s(f)|$, for the EPA's used in (a) and (b).

Fig. 11. Influence of variations in the choice of the residual earcanal volume $V_C$: (a) Effects on the magnitude and angle of the admittance at the eardrum, $|Y_d(f)|$ in dB re 1 mS (cgs), above, and $\measuredangle Y_d(f)$ in degrees, below; effects on the sound-pressure transformation of the earcanal, $|P_d(f)/P_{ec}(f)|$ in dB, middle. (b) Effects on the real and imaginary components of the impedance at the eardrum, $R_d(f)$ above, $X_d(f)$ below, both in cgs Ω. Values of the admittance at the EPA-tip, $Y_e(f)$, shown in (a) and the components $R_e(f)$ and $X_e(f)$ shown in (b). See text for details.

Fig. 12. Eardrum admittance $Y_d(f)$ for each subject. Results shown are the averages of the intrasession mean $Y_d(f)$'s. Admittance magnitude, $|Y_d(f)|$ in dB re 1 mS (cgs), above; admittance angle,

$\measuredangle Y_d(f)$ in degrees, below. Also shown are the standard deviations of the intersession averages, $\sigma[Y_d(f)]$ in dB and degrees. Note that the ordinates for $\sigma[Y_d(f)]$ are expanded re that for $Y_d(f)$.

Fig. 13. Group-average eardrum admittance, $Y_d(f)$ for $\bar{S}$, obtained from the individual-subject results in Fig. 12. Admittance values for ±1 intersubject standard deviation about the group-average $Y_d(f)$ also shown. Admittance magnitude, $|Y_d(f)|$ in dB re 1 mS (cgs), above; admittance angle, $\measuredangle Y_d(f)$ in degrees, below. Calculated values of the sound-pressure transformation of the earcanal, $|P_d(f)/P_{ec}(f)|$ in dB, middle.

Fig. 14. Individual-subject results for the real and imaginary components of the impedance at the eardrum, $Z_d(f) = R_d(f) + jX_d(f) = 1/Y_d(f)$. Results calculated from the $Y_d(f)$'s in Fig. 12. Resistance, $R_d(f)$ above, and reactance, $X_d(f)$ below, both in cgs Ω.

Fig. 15. Group-average results for the resistance at the eardrum, $R_d(f)$ above, and the reactance at the eardrum, $X_d(f)$ below, both in cgs Ω. Results calculated from the admittance for $\bar{S}$ given in Fig. 13. Also shown are the "limits" of the individual-subject results given in Fig. 14.

Fig. 16. Individual-subject results for the sound-pressure transformation of the earcanal, $|P_d(f)/P_{ec}(f)|$ in dB, calculated from the $Y_d(f)$'s in Fig. 12.

Fig. 17. Standard deviations of the admittance measurements: Intersubject $\sigma[Y_d(f)]$ calculated from the individual-subject $Y_d(f)$'s in Fig. 12. [This $\sigma$ is the same as that shown in Fig. 13.] Also shown are the minimum and maximum intrasubject $\sigma[Y_d(f)]$'s obtained as the "limits" of the $\sigma[Y_d(f)]$'s given in Fig. 12 Magnitude, $\sigma[|Y_d(f)|]$ in dB, above; angle, $\sigma[\measuredangle Y_d(f)]$ in degrees, below.

Fig. 18. Intersession $\sigma[Y_d(f)]$'s for each subject replotted from Fig. 12. Also shown are the maxima and minima of the intrasession $\sigma[Y_d(f)]$'s obtained for each subject. Magnitude, $\sigma[|Y_d(f)|]$ in dB, above; angle, $\sigma[\measuredangle Y_d(f)]$ in degrees, below.

Fig. 19. Eardrum admittance $Y_d(f)$ from an experiment on subject S1 using a hard eartip as compared to the $Y_d(f)$ for S1 determined from tests using the inflatable cuff (replotted from Fig. 12). Admittance magnitude, $|Y_d(f)|$ in dB re 1 mS (cgs), above, admittance angle, $\measuredangle Y_d(f)$ in degrees, below.

Fig. 20. Comparison of our average results for $Y_d(f)$ to values reported in other studies: Average results for 5 subjects from Blauert and Platte (1974, Fig. 4) and for 10 subjects from Møller (1960, Fig. 7). Median results for 10 males from Zwislocki (1970, Figs. 18 and 19). Results for $|Y_d(f)|$ for 1 cadaver from Fischler et al (1966, Fig. 4). (a) Eardrum impedance components of resistance, $R_d(f)$ above, and reactance, $X_d(f)$ below, both in cgs $\Omega$. (b) Admittance magnitude, $|Y_d(f)|$ in dB re 1 mS (cgs) above;

admittance angle, $\measuredangle Y_d(f)$ in degrees, below. Calculated values of $|P_d(f)/P_{ec}(f)|$, in dB, middle.

Fig. 21. Comparison of our average values for $Y_d(f)$ to values suggested by Shaw (1975b, Fig. 11). Proceeding from the top down: admittance magnitude, $|Y_d(f)|$ in dB re 1 mS (cgs); admittance angle, $\measuredangle Y_d(f)$ in degrees; resistance, $R_d(f)$ in cgs Ω; reactance, $X_d(f)$ in cgs Ω; values of $|P_d(f)/P_{ec}(f)|$, in dB, calculated from $Y_d(f)$.

Fig. 22. Comparison of our results for $R_d(f)$ and $X_d(f)$ to values reported on normal ears in two "clinical" studies in which the Zwislocki acoustic bridge was used. Results obtained on 25 subjects, aged 19 to 29 and predominantly male, from Burke, Nilges, and Henry (1970, Table 2), and on 33 subjects from Zwislocki and Feldman (1970, values reported by Lilly, 1973, Table III). Resistance, $R_d(f)$ above, and reactance $X_d(f)$ below, both in cgs Ω.

Fig. 23. Comparison of our average results for $Y_d(f)$ to values calculated from the middle-ear model of Zwislocki (1962). [Fig. 17 of Part 2 shows the Zwislocki model.] Proceeding from the top down: admittance magnitude, $|Y_d(f)|$ in dB re 1 mS (cgs); admittance angle, $\measuredangle Y_d(f)$ in degrees; resistance, $R_d(f)$ in cgs Ω; reactance, $X_d(f)$ in cgs Ω; values of $|P_d(f)/P_{ec}(f)|$, in dB, calculated from $Y_d(f)$.

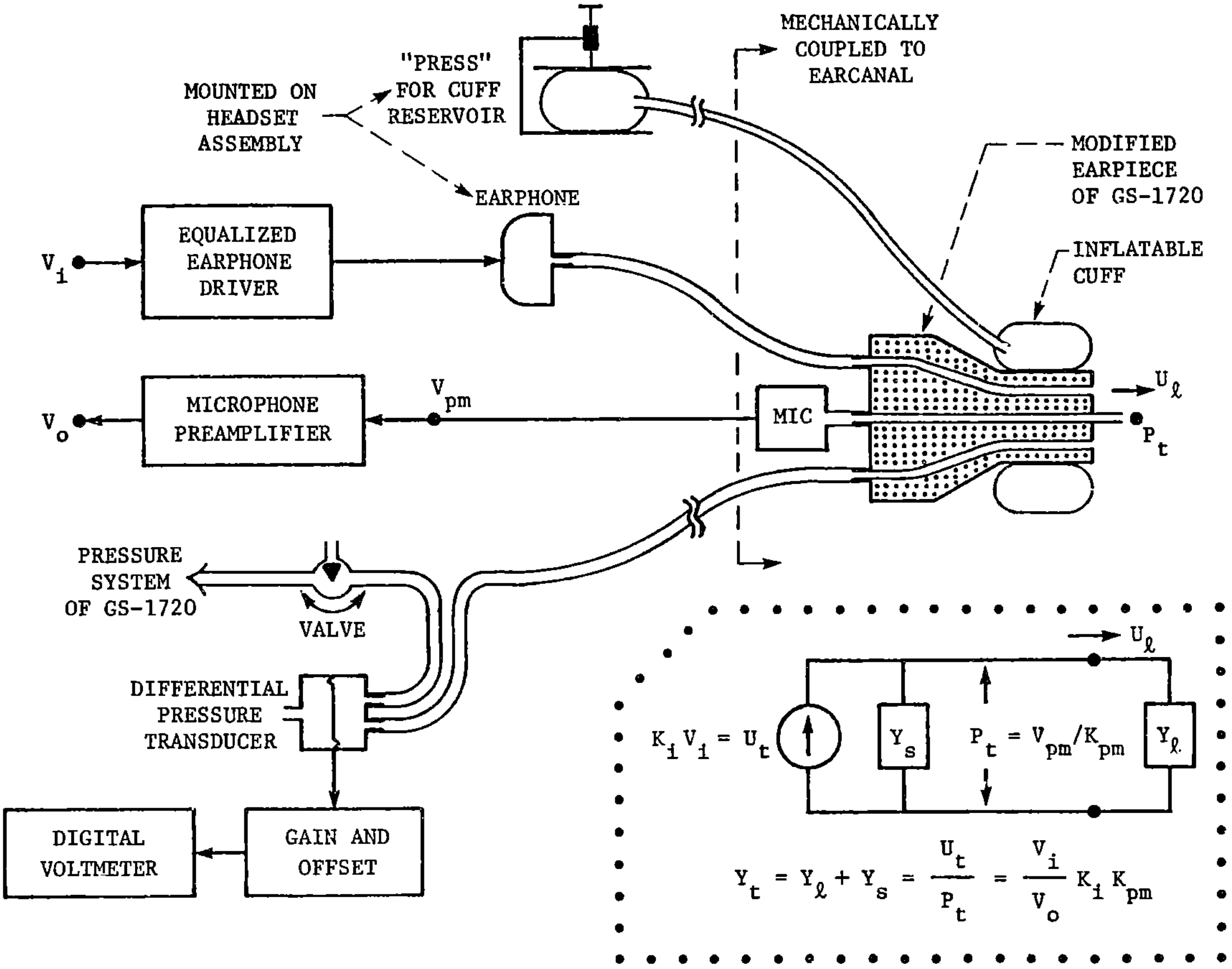

MOUNTED ON HEADSET ASSEMBLY
"PRESS" FOR CUFF RESERVOIR
EARPHONE
MECHANICALLY COUPLED TO EARCANAL
MODIFIED EARPIECE OF GS-1720
INFLATABLE CUFF
$V_i$
EQUALIZED EARPHONE DRIVER
$V_o$
MICROPHONE PREAMPLIFIER
$V_{pm}$
MIC
$U_\ell$
$P_t$
PRESSURE SYSTEM OF GS-1720
VALVE
DIFFERENTIAL PRESSURE TRANSDUCER
DIGITAL VOLTMETER
GAIN AND OFFSET
$U_\ell$
$K_i V_i = U_t$
$Y_s$
$P_t = V_{pm}/K_{pm}$
$Y_\ell$
$Y_t = Y_\ell + Y_s = \frac{U_t}{P_t} = \frac{V_i}{V_o} K_i K_{pm}$

FIGURE 1

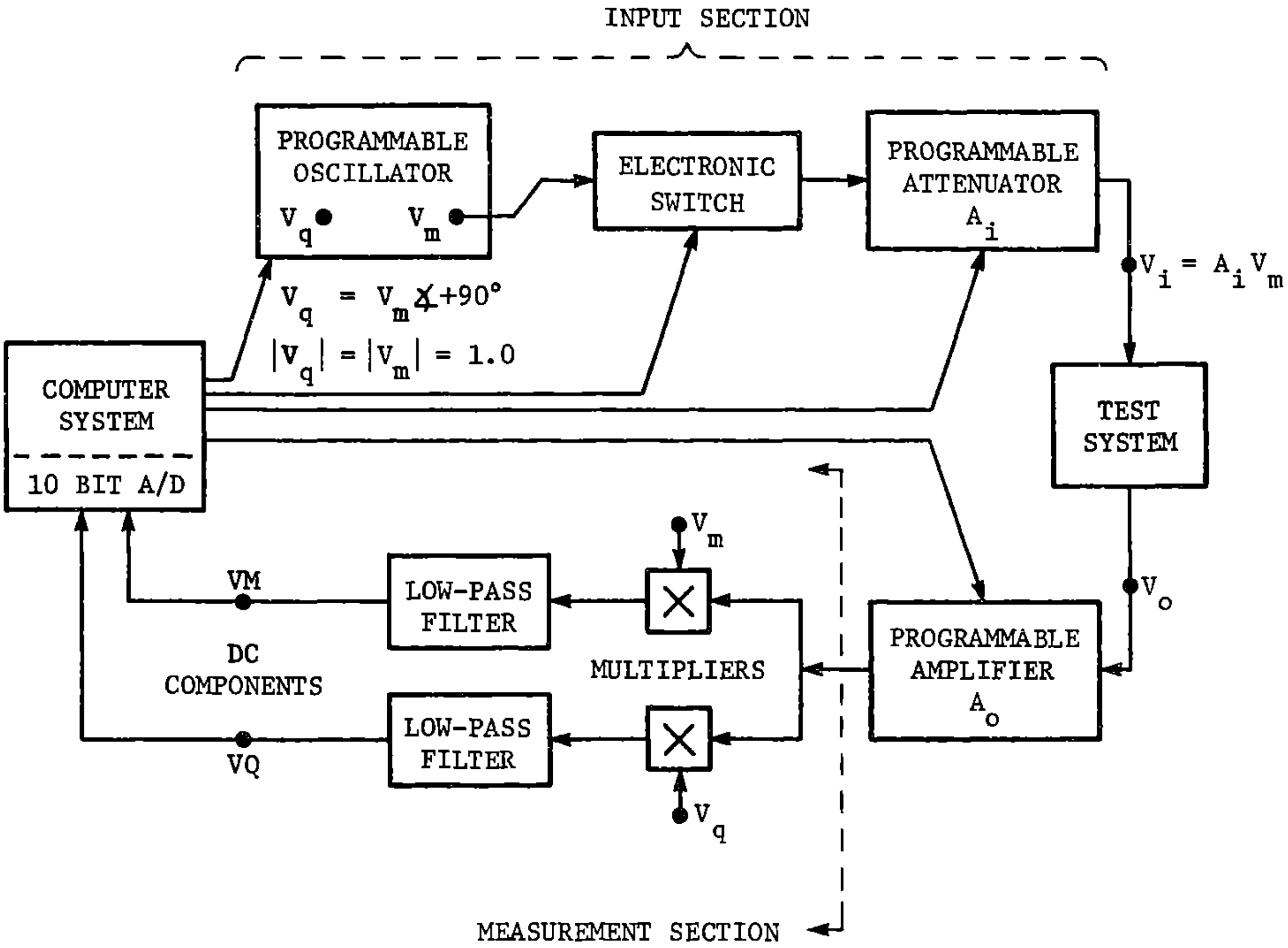

INPUT SECTION
PROGRAMMABLE OSCILLATOR
$V_q$
$V_m$
ELECTRONIC SWITCH
PROGRAMMABLE ATTENUATOR $A_i$
$V_i = A_i V_m$
$V_q = V_m \measuredangle +90°$
$|V_q| = |V_m| = 1.0$
COMPUTER SYSTEM
10 BIT A/D
TEST SYSTEM
$V_o$
$V_m$
VM
LOW-PASS FILTER
DC COMPONENTS
MULTIPLIERS
PROGRAMMABLE AMPLIFIER $A_o$
LOW-PASS FILTER
VQ
$V_q$
MEASUREMENT SECTION

FIGURE 2

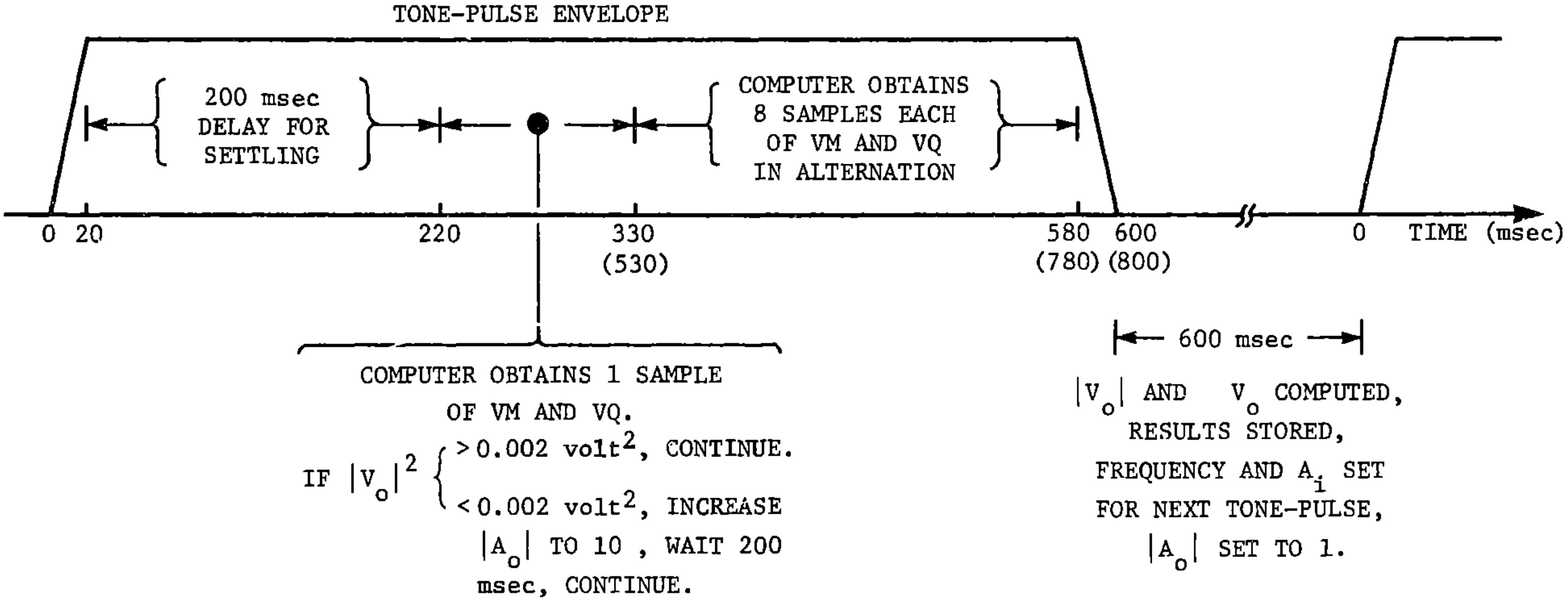

TONE-PULSE ENVELOPE
200 msec DELAY FOR SETTLING
COMPUTER OBTAINS 8 SAMPLES EACH OF VM AND VQ IN ALTERNATION
0
20
220
330 (530)
580 (780)
600 (800)
0
TIME (msec)
COMPUTER OBTAINS 1 SAMPLE OF VM AND VQ.
IF $|V_o|^2$ > 0.002 volt$^2$, CONTINUE.
< 0.002 volt$^2$, INCREASE $|A_o|$ TO 10, WAIT 200 msec, CONTINUE.
600 msec
$|V_o|$ AND $V_o$ COMPUTED, RESULTS STORED, FREQUENCY AND $A_i$ SET FOR NEXT TONE-PULSE, $|A_o|$ SET TO 1.

FIGURE 3

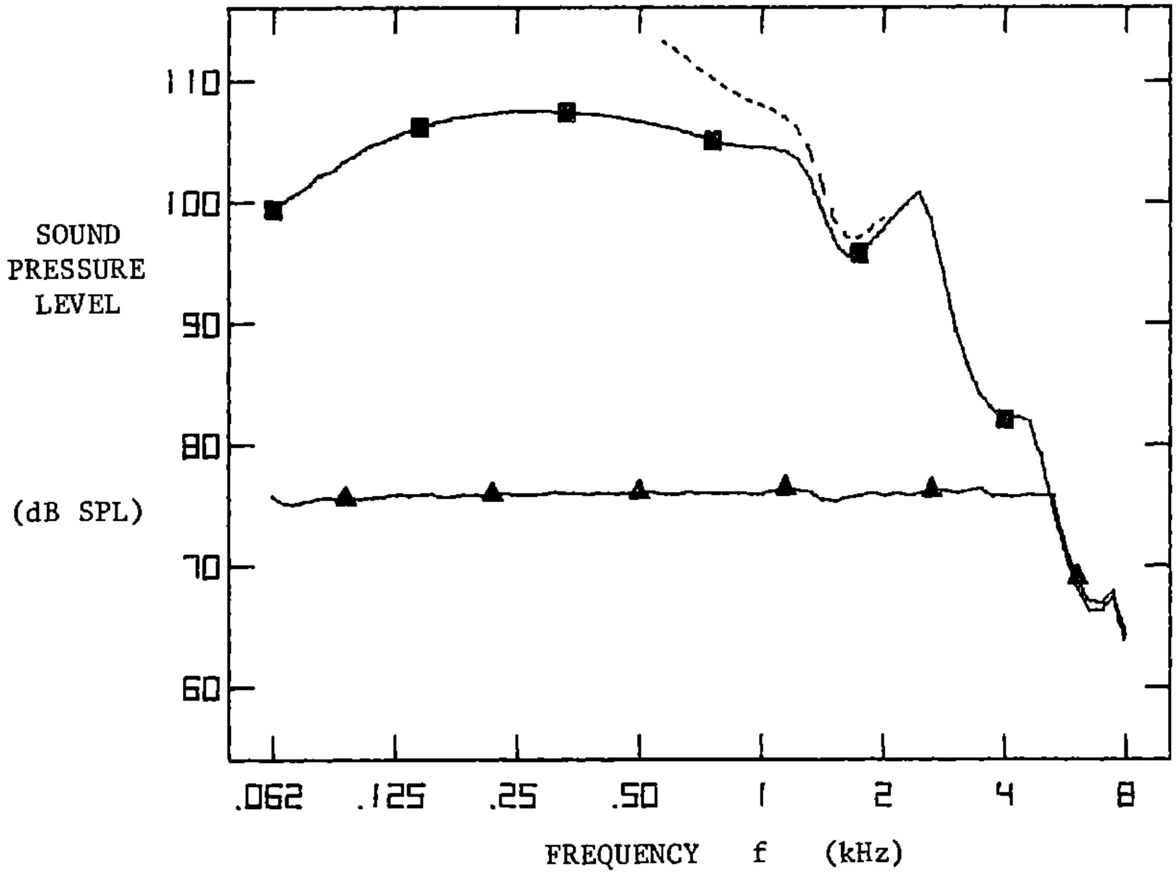

110
100
90
80
70
60
SOUND PRESSURE LEVEL
(dB SPL)
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 4

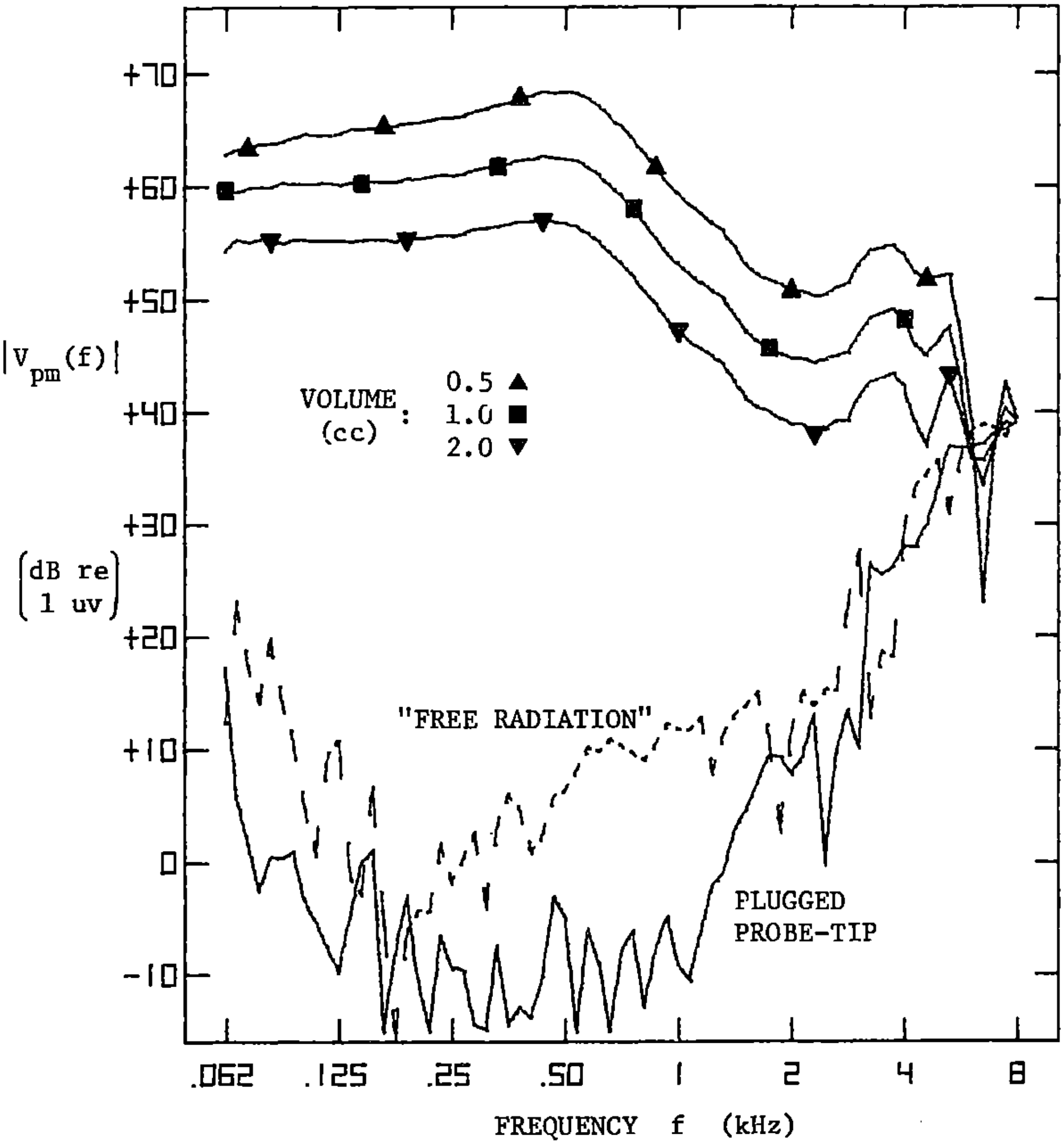

+70
+60
+50
+40
+30
+20
+10
0
-10
|V_pm(f)|
(dB re 1 uv)
VOLUME (cc): 0.5 ▲ 1.0 ■ 2.0 ▼
"FREE RADIATION"
PLUGGED PROBE-TIP
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 5

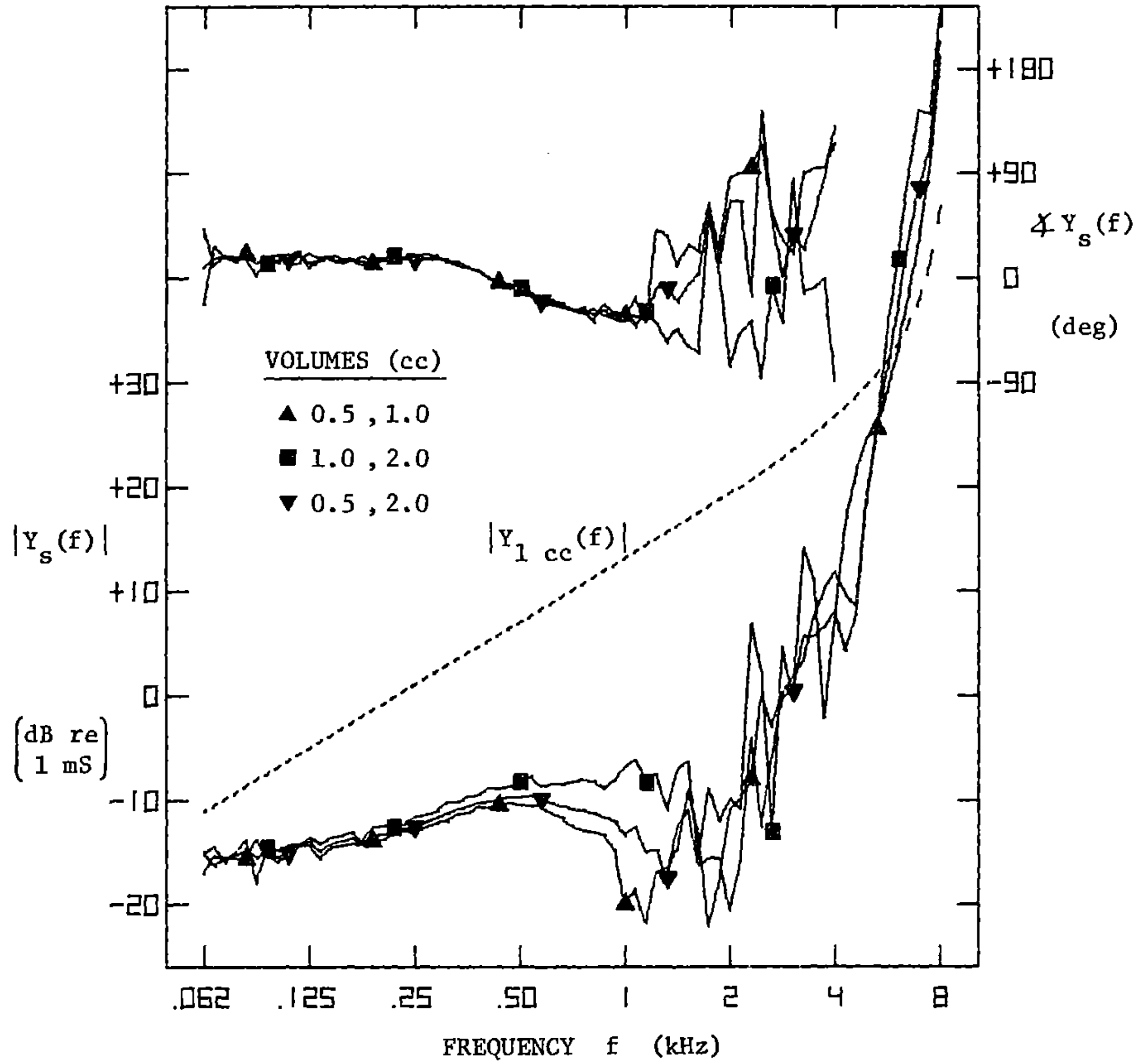

VOLUMES (cc)
▲ 0.5 , 1.0
■ 1.0 , 2.0
▼ 0.5 , 2.0
$|Y_{1\ cc}(f)|$
$|Y_s(f)|$
$\left(\begin{matrix}\text{dB re}\\ \text{1 mS}\end{matrix}\right)$
+30
+20
+10
0
-10
-20
+180
+90
0
-90
$\measuredangle Y_s(f)$
(deg)
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 6

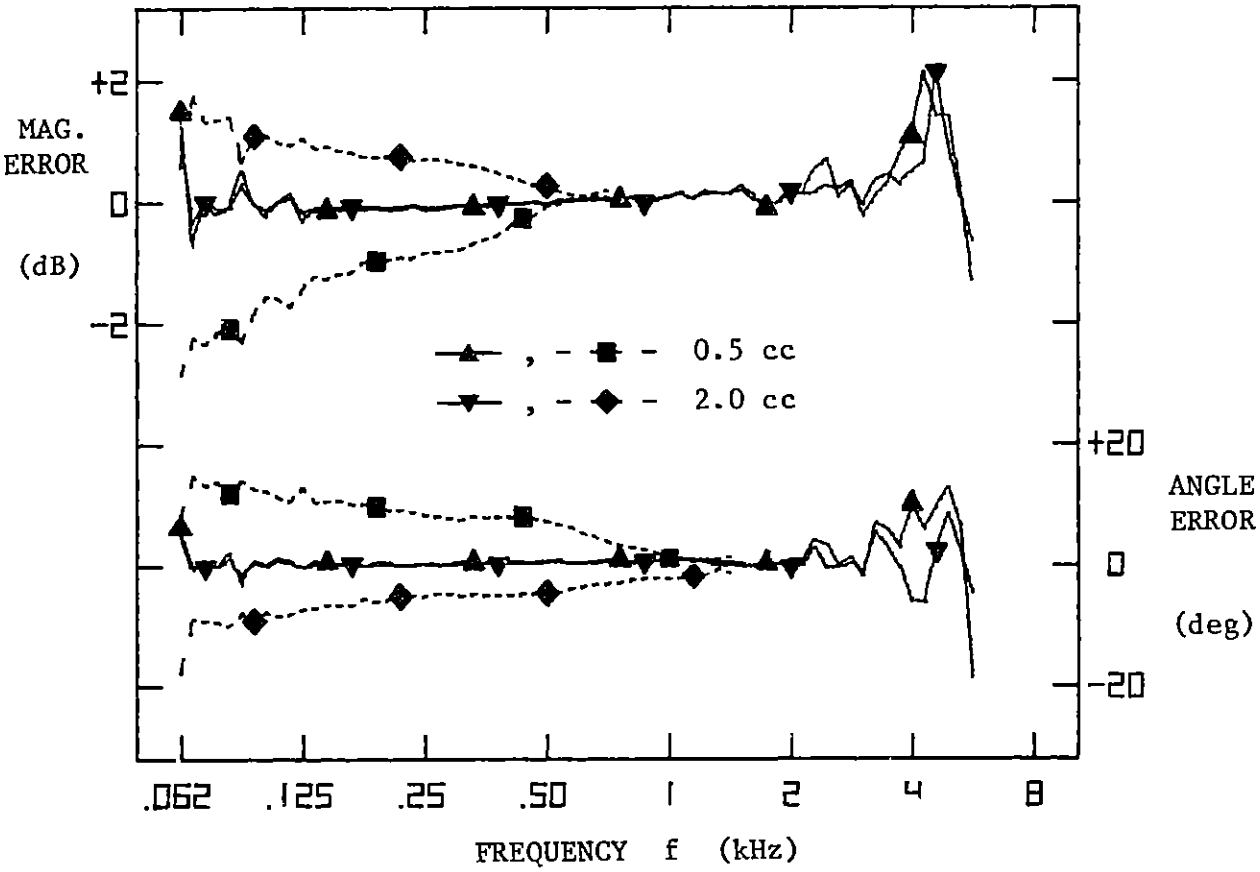

+2
MAG.
ERROR
0
(dB)
-2
—▲— , - ■ - 0.5 cc
—▼— , - ◆ - 2.0 cc
+20
ANGLE
ERROR
0
(deg)
-20
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 7

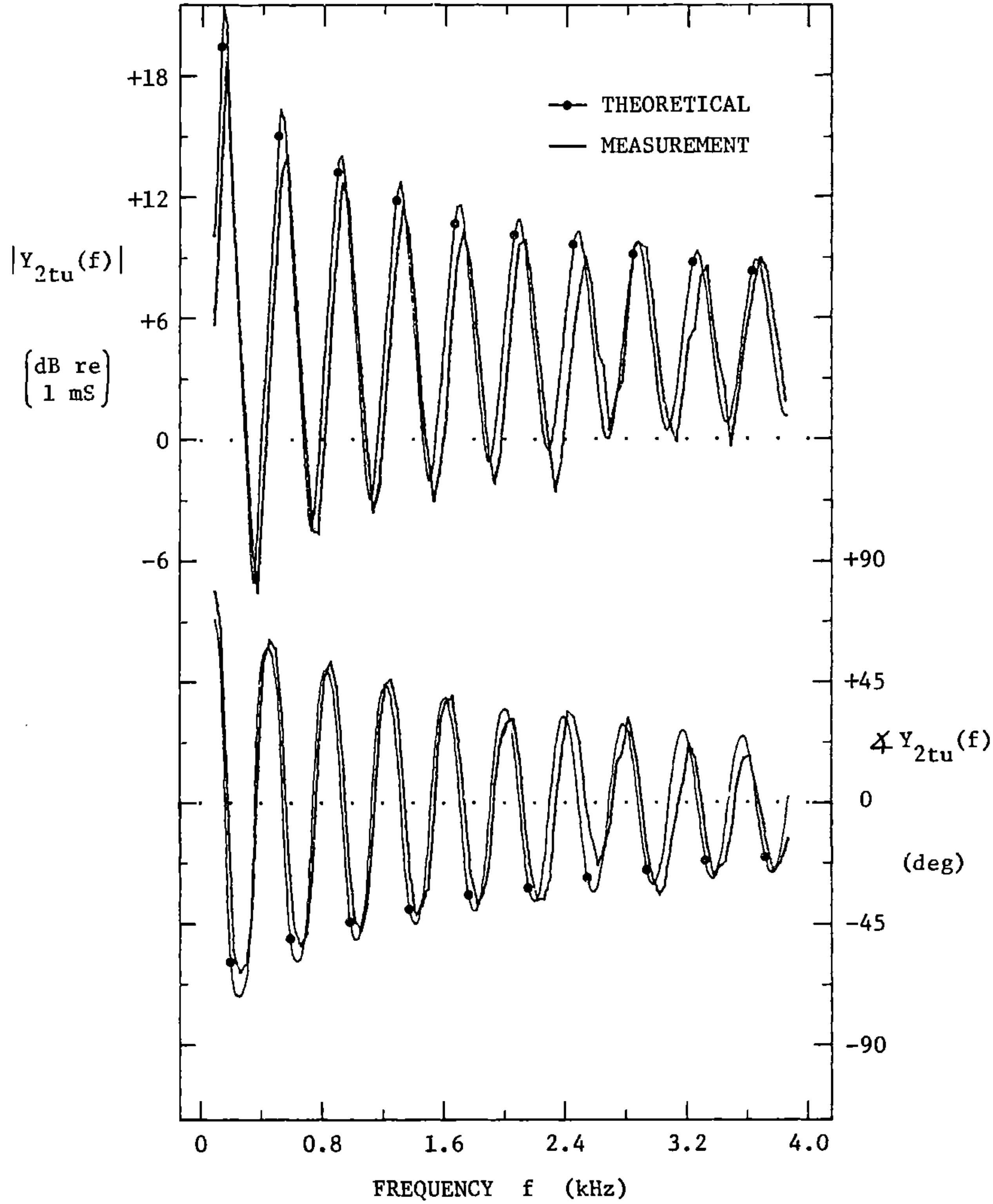

+18
+12
+6
0
-6
|$Y_{2tu}(f)$|
(dB re 1 mS)
THEORETICAL
MEASUREMENT
+90
+45
0
-45
-90
∡ $Y_{2tu}(f)$
(deg)
0
0.8
1.6
2.4
3.2
4.0
FREQUENCY f (kHz)

FIGURE 8

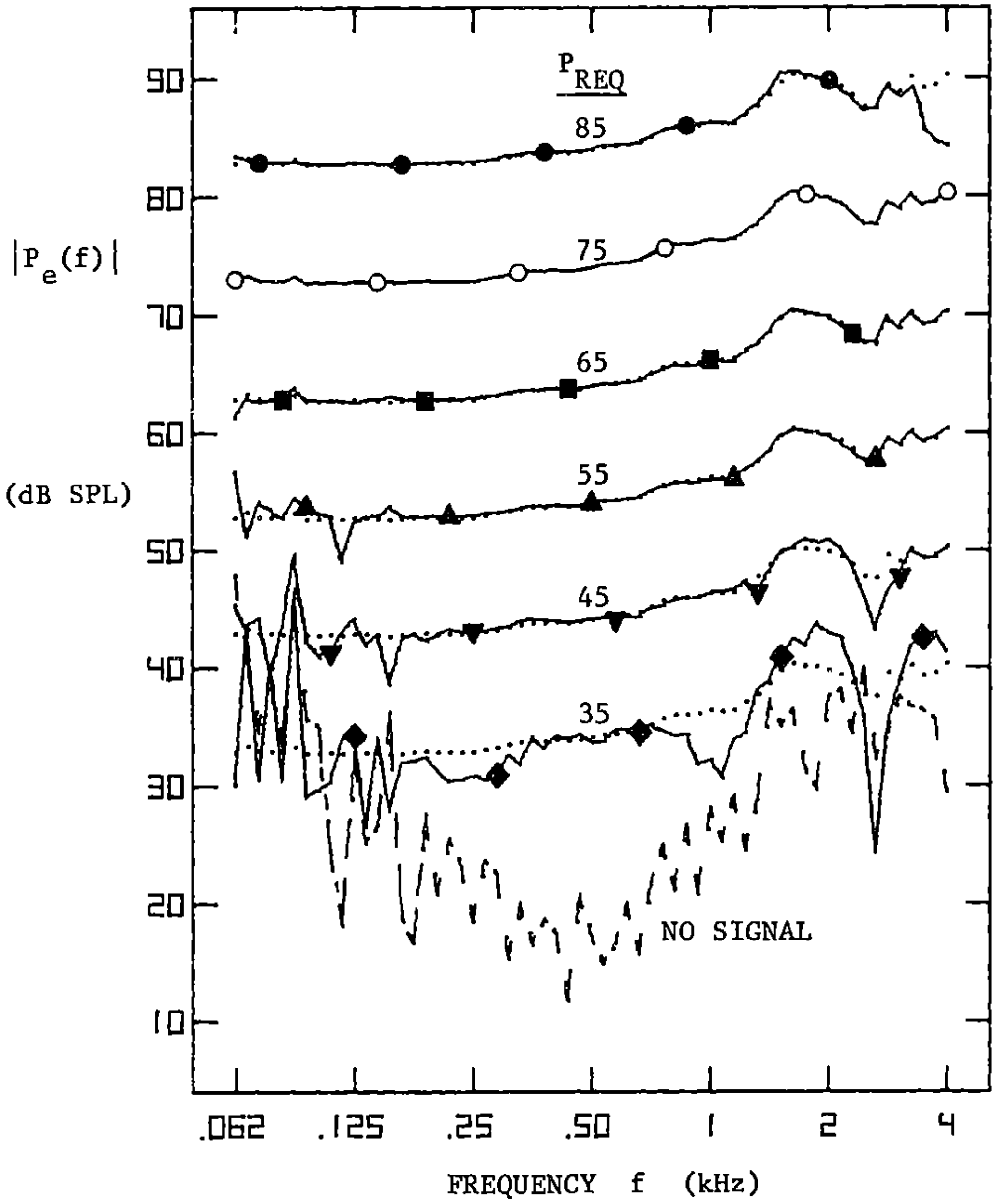
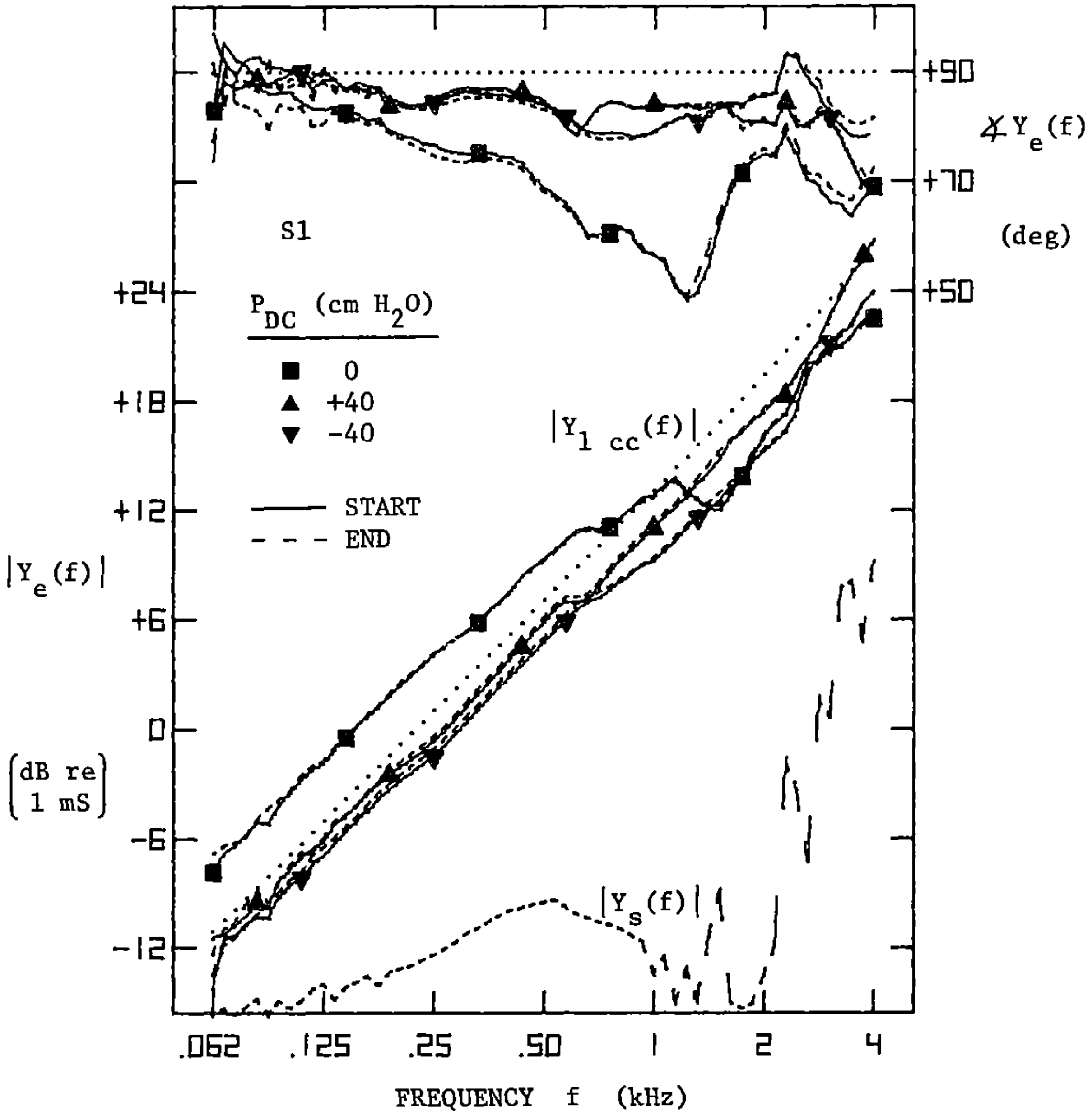
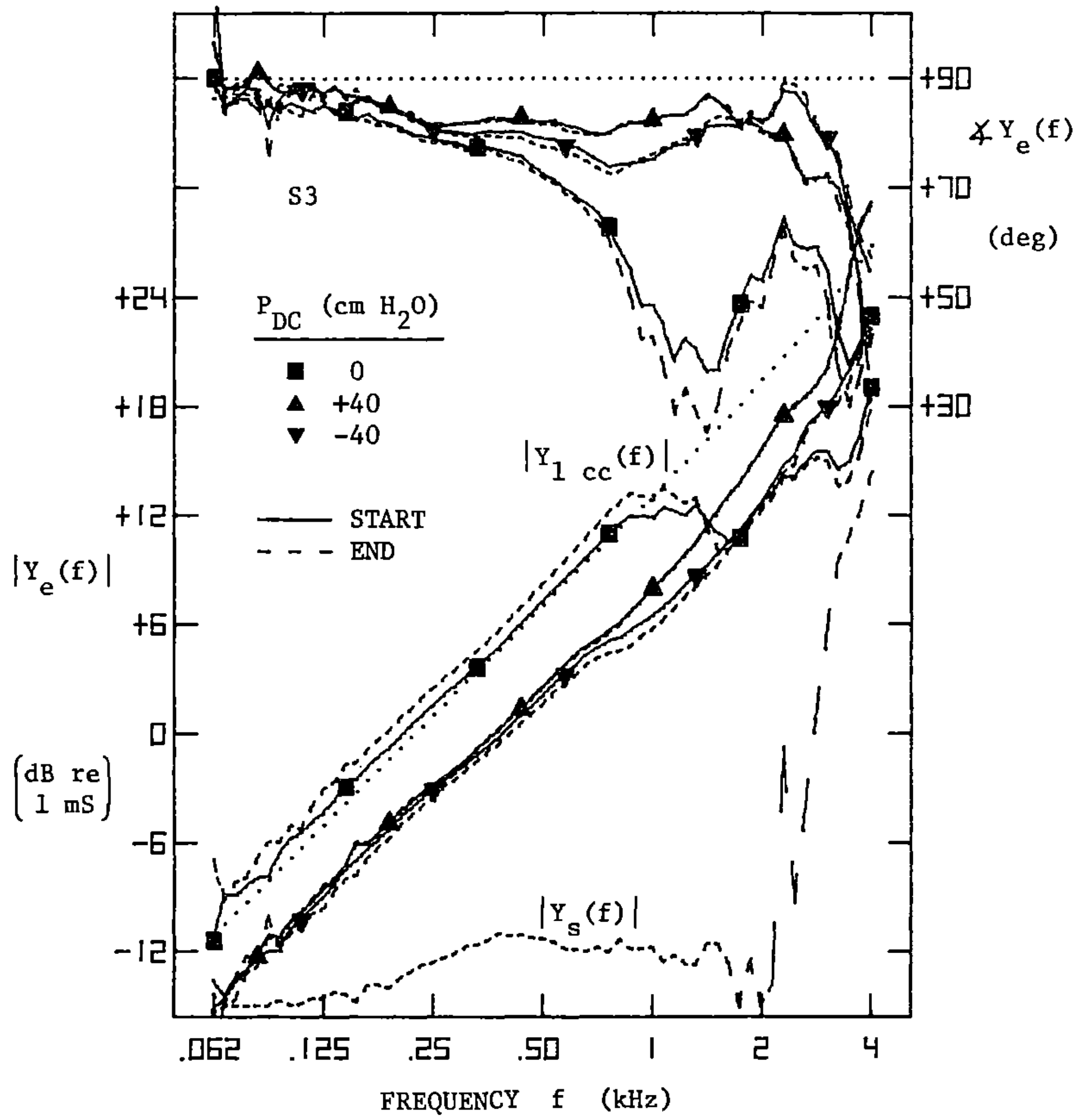

$|P_e(f)|$
(dB SPL)
$P_{REQ}$
85
75
65
55
45
35
NO SIGNAL
90
80
70
60
50
40
30
20
10
.062
.125
.25
.50
1
2
4
FREQUENCY f (kHz)

FIGURE 9



+90
∡Y_e(f)
+70
(deg)
+50
S1
+24
+18
+12
+6
0
-6
-12
|Y_e(f)|
(dB re 1 mS)
P_DC (cm H_2O)
■ 0
▲ +40
▼ -40
—— START
- - - END
|Y_1 cc(f)|
|Y_s(f)|
.062
.125
.25
.50
1
2
4
FREQUENCY f (kHz)

FIGURE 10(a)



S3
$P_{DC}$ (cm $H_2O$)
■ 0
▲ +40
▼ -40
START
END
$|Y_{1\ cc}(f)|$
$|Y_s(f)|$
$|Y_e(f)|$
(dB re 1 mS)
+24
+18
+12
+6
0
-6
-12
$\measuredangle Y_e(f)$
(deg)
+90
+70
+50
+30
.062
.125
.25
.50
1
2
4
FREQUENCY f (kHz)

FIGURE 10(b)

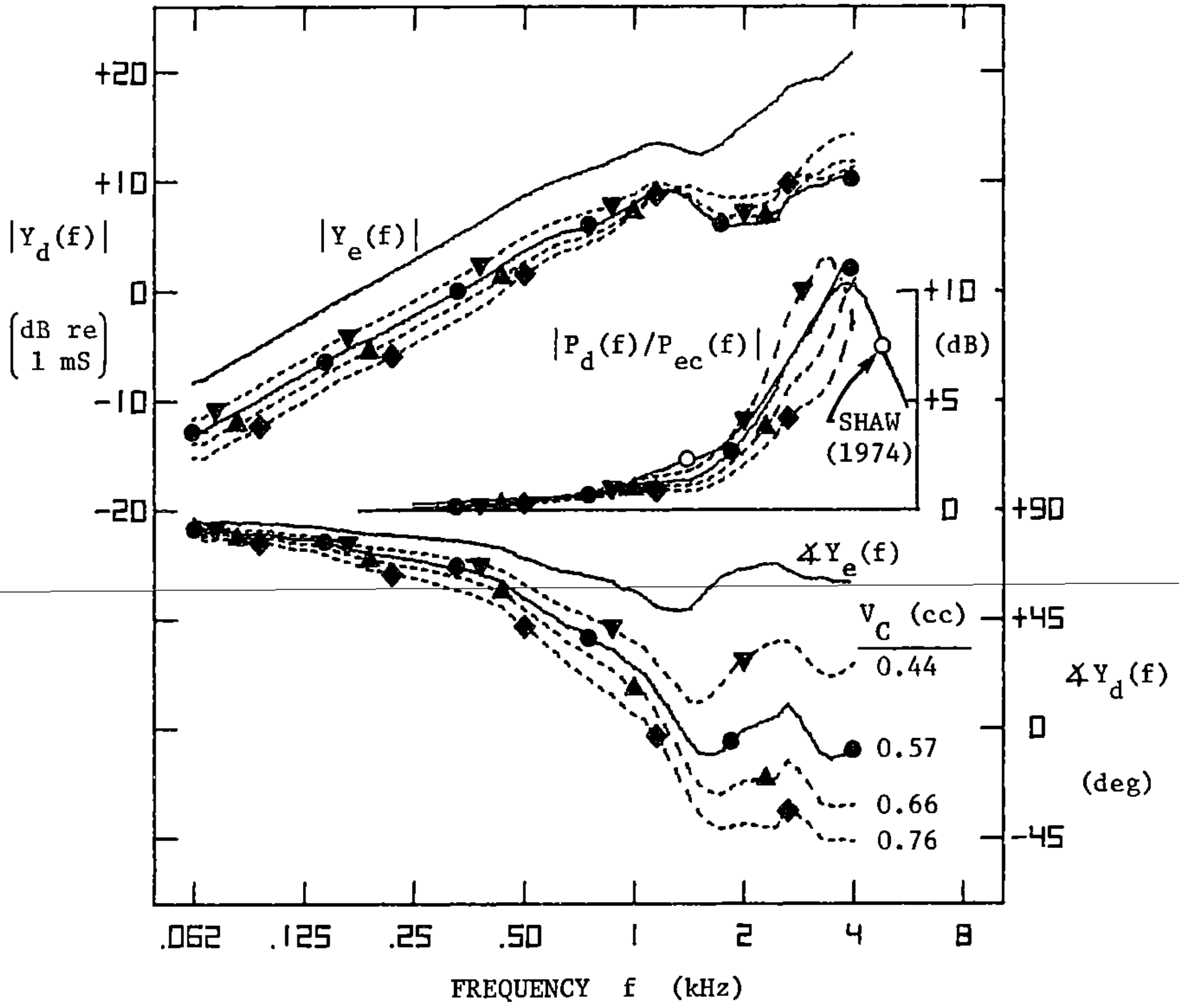

+20
+10
0
-10
-20
|Y_d(f)|
(dB re 1 mS)
|Y_e(f)|
|P_d(f)/P_ec(f)|
+10
+5
0
(dB)
SHAW (1974)
+90
+45
0
-45
∡Y_e(f)
∡Y_d(f)
(deg)
V_C (cc)
0.44
0.57
0.66
0.76
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 11 (a)

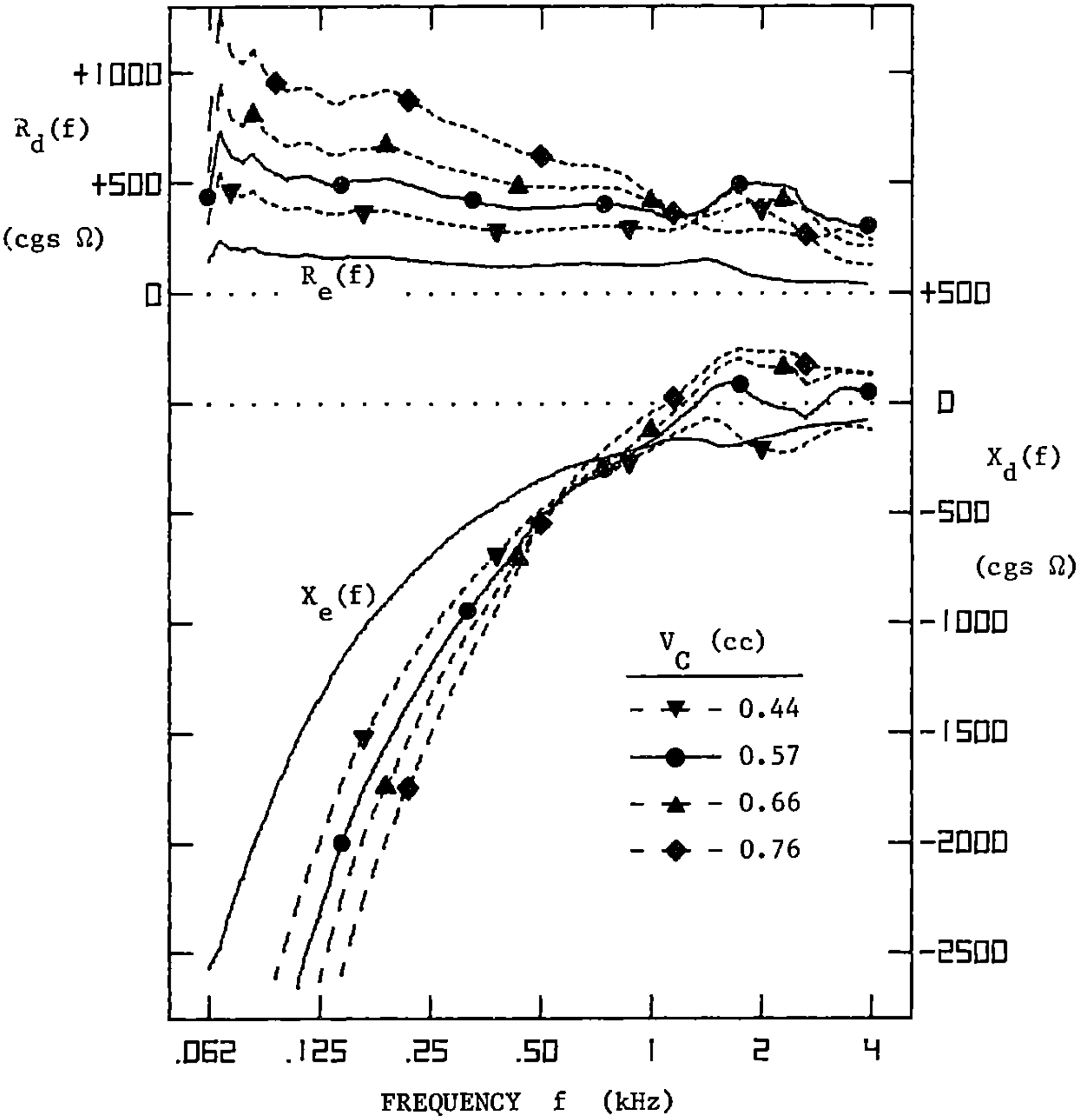

+1000
$R_d(f)$
+500
(cgs Ω)
0
$R_e(f)$
+500
0
$X_d(f)$
-500
(cgs Ω)
-1000
-1500
-2000
-2500
$X_e(f)$
$V_C$ (cc)
▼ 0.44
● 0.57
▲ 0.66
◆ 0.76
.062
.125
.25
.50
1
2
4
FREQUENCY f (kHz)

FIGURE 11 (b)

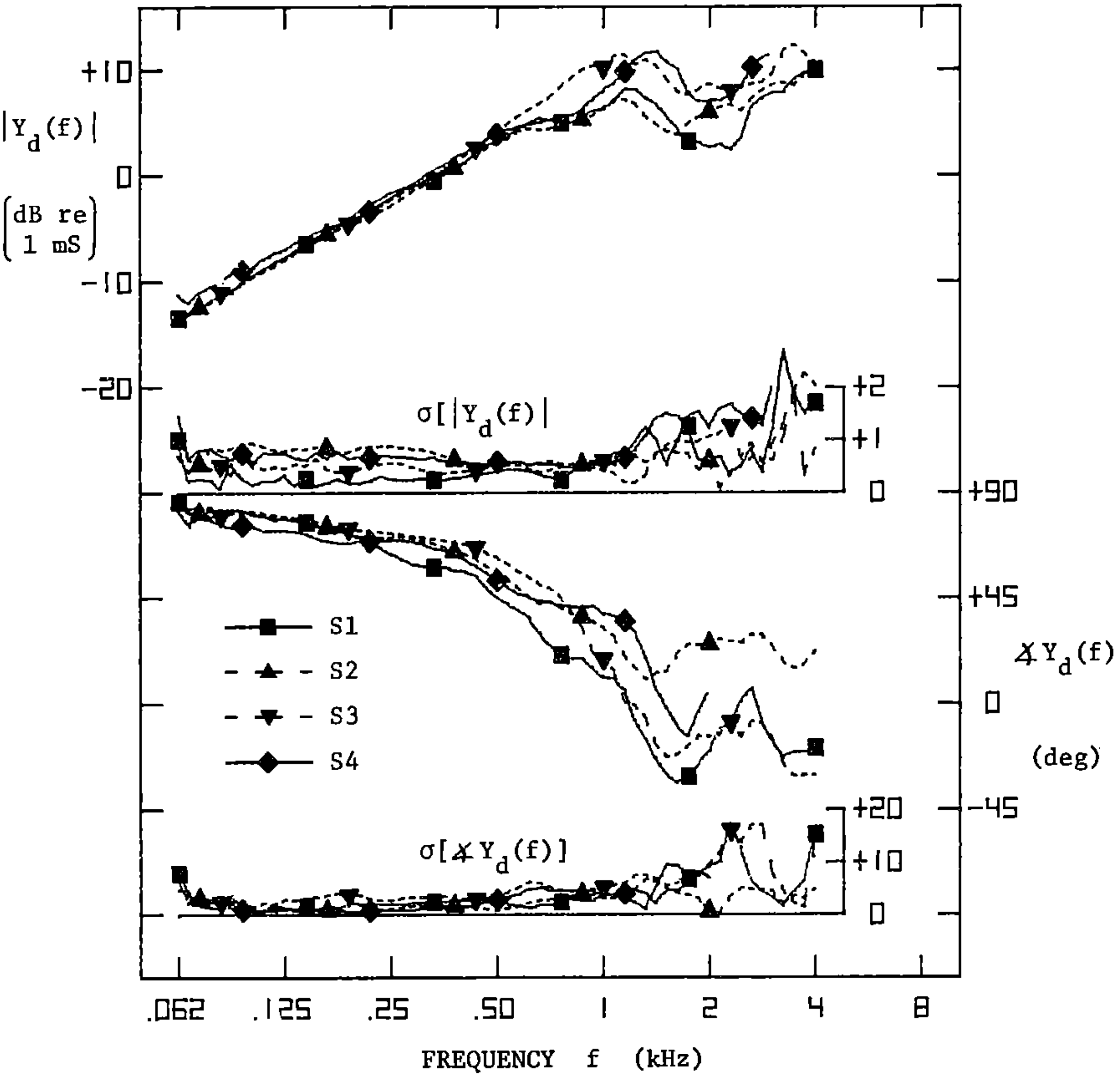
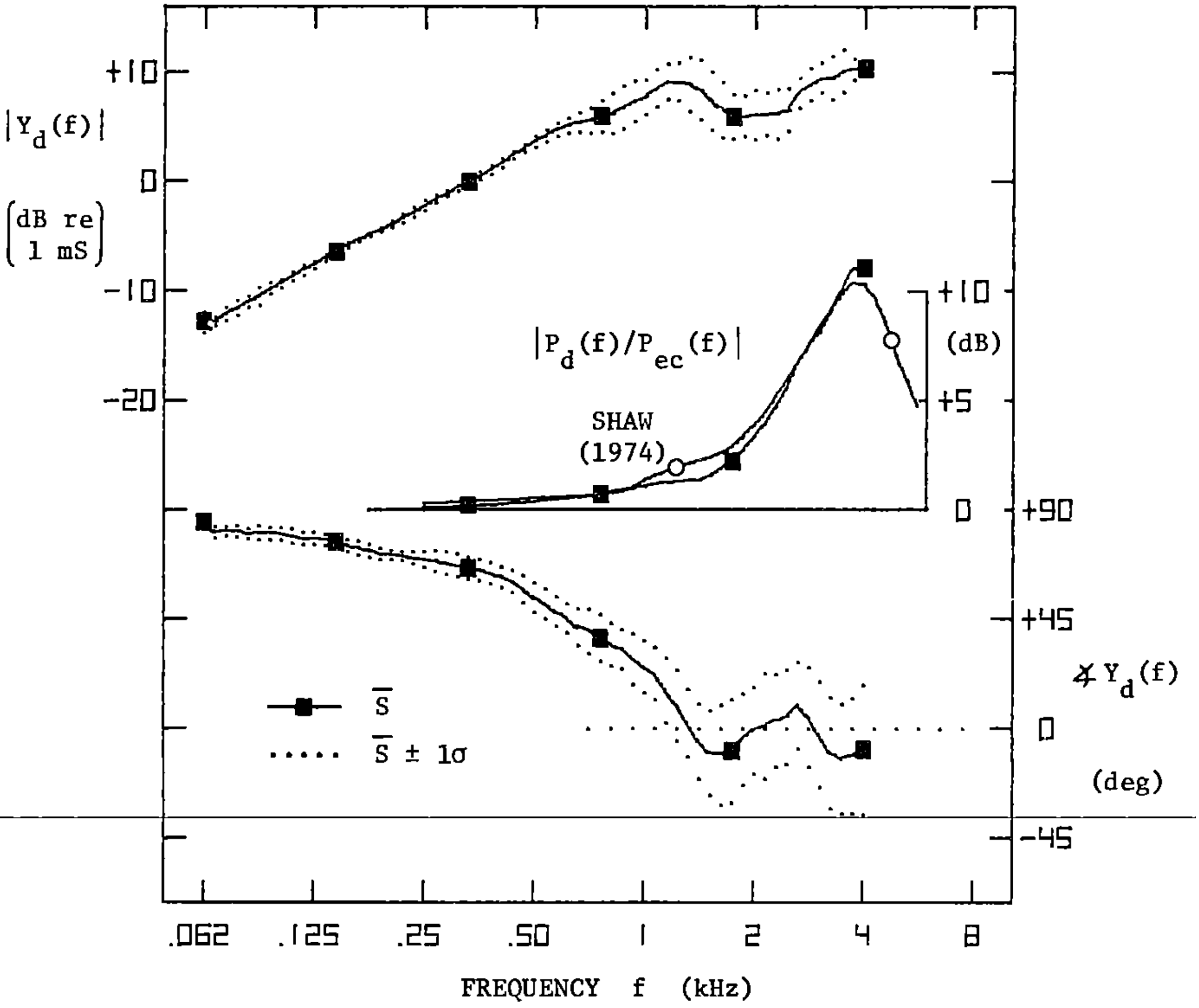

+10
$|Y_d(f)|$
0
(dB re 1 mS)
-10
-20
$\sigma[|Y_d(f)|$
+2
+1
0
+90
+45
$\measuredangle Y_d(f)$
0
(deg)
-45
S1
S2
S3
S4
$\sigma[\measuredangle Y_d(f)]$
+20
+10
0
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 12



+10
0
-10
-20
$|Y_d(f)|$
(dB re 1 mS)
$|P_d(f)/P_{ec}(f)|$
+10
(dB)
+5
0
SHAW (1974)
+90
+45
0
-45
$\measuredangle Y_d(f)$
(deg)
$\overline{S}$
$\overline{S} \pm 1\sigma$
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 13

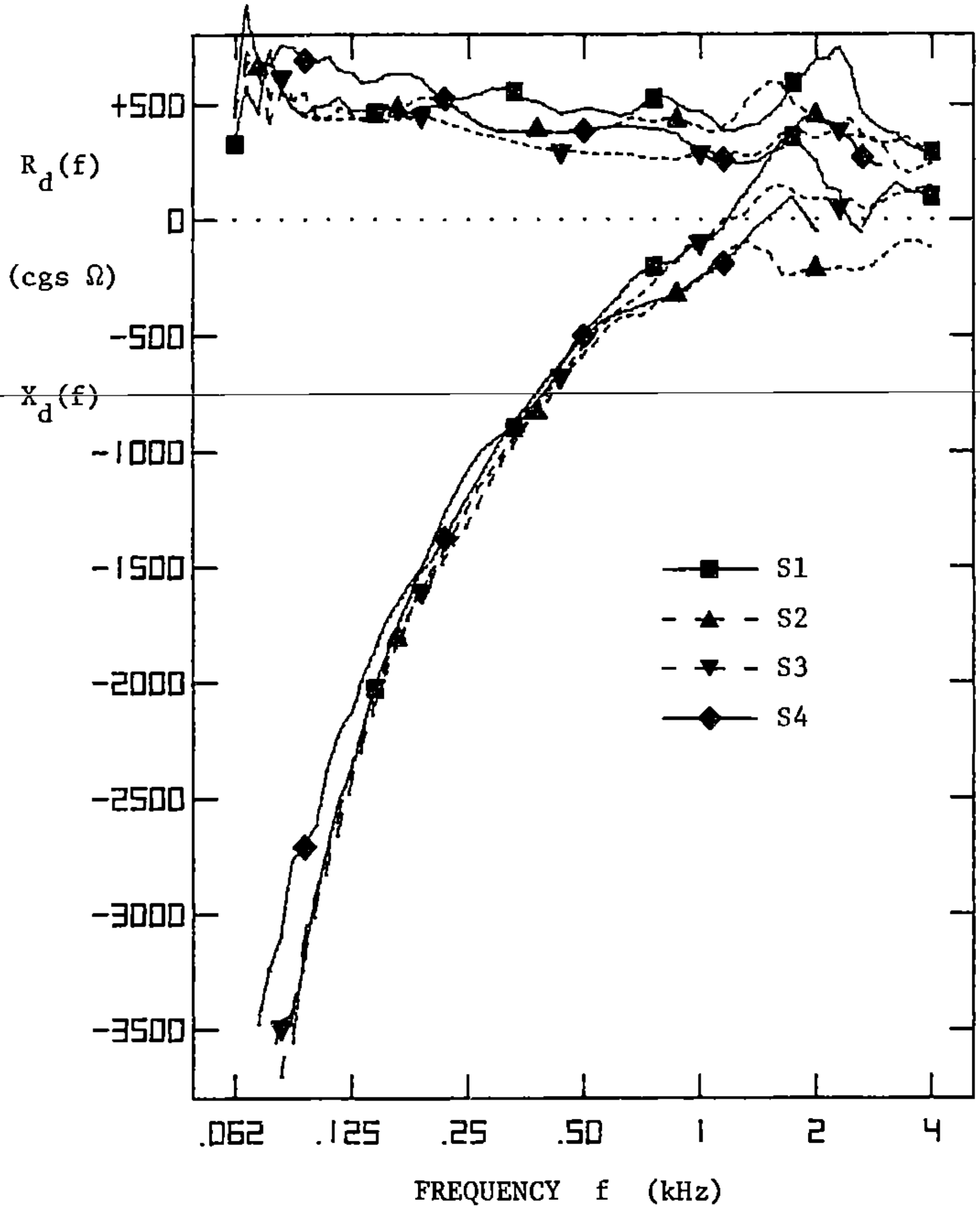

+500
$R_d(f)$
0
(cgs Ω)
-500
$X_d(f)$
-1000
-1500
-2000
-2500
-3000
-3500
.062
.125
.25
.50
1
2
4
FREQUENCY f (kHz)
S1
S2
S3
S4

FIGURE 14

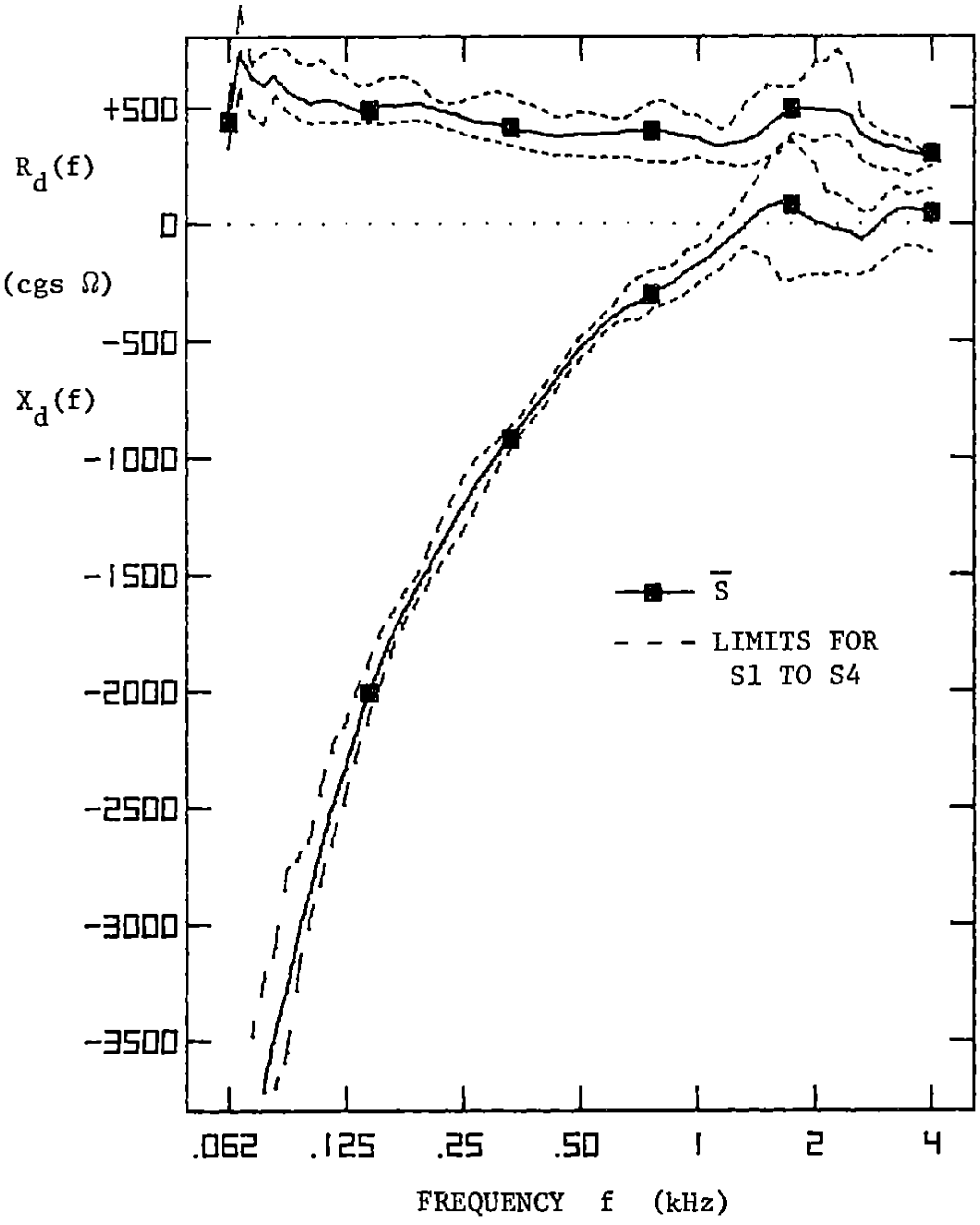

+500
$R_d(f)$
0
(cgs Ω)
-500
$X_d(f)$
-1000
-1500
-2000
-2500
-3000
-3500
$\overline{S}$
LIMITS FOR S1 TO S4
.062
.125
.25
.50
1
2
4
FREQUENCY f (kHz)

FIGURE 15

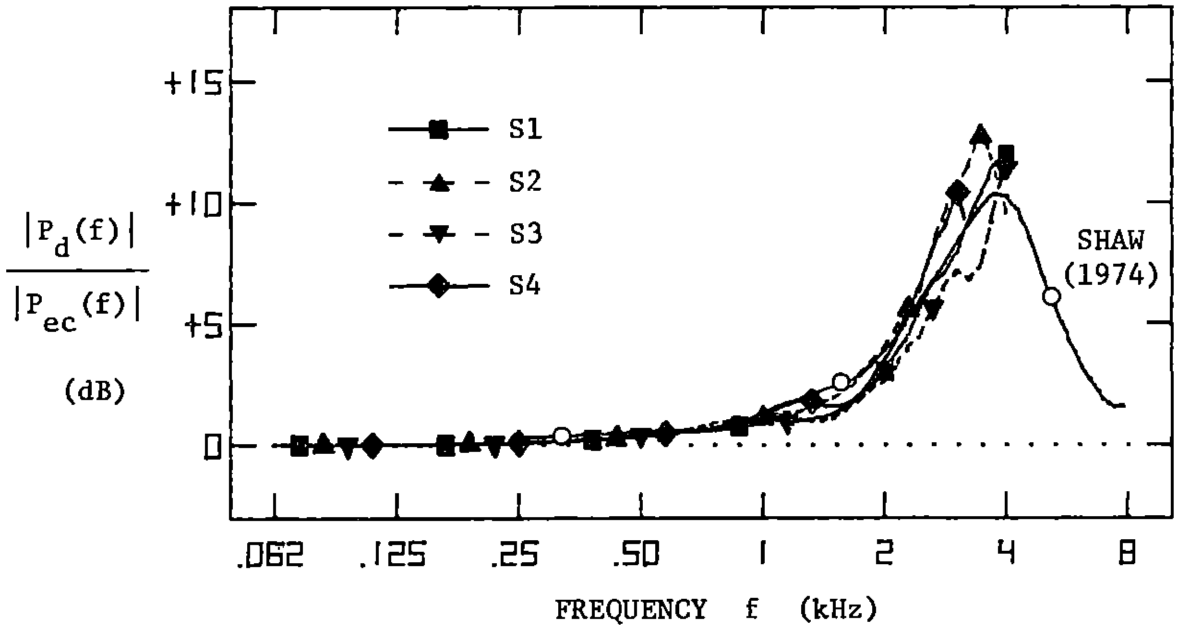

+15
+10
+5
0
$\frac{|P_d(f)|}{|P_{ec}(f)|}$
(dB)
S1
S2
S3
S4
SHAW
(1974)
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 16

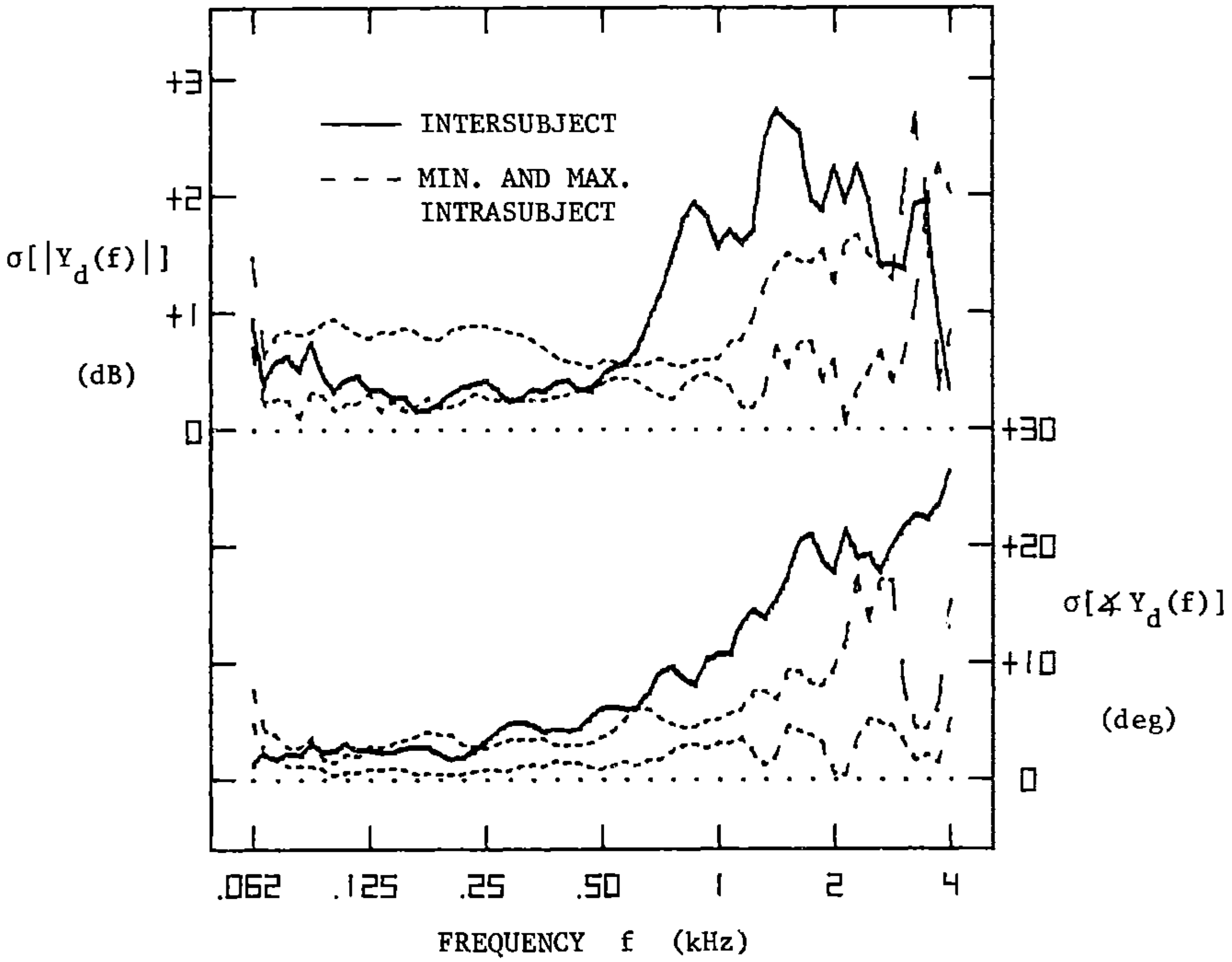

+3
+2
+1
0
$\sigma[|Y_d(f)|]$
(dB)
INTERSUBJECT
MIN. AND MAX. INTRASUBJECT
+30
+20
+10
0
$\sigma[\measuredangle Y_d(f)]$
(deg)
.062
.125
.25
.50
1
2
4
FREQUENCY f (kHz)

FIGURE 17

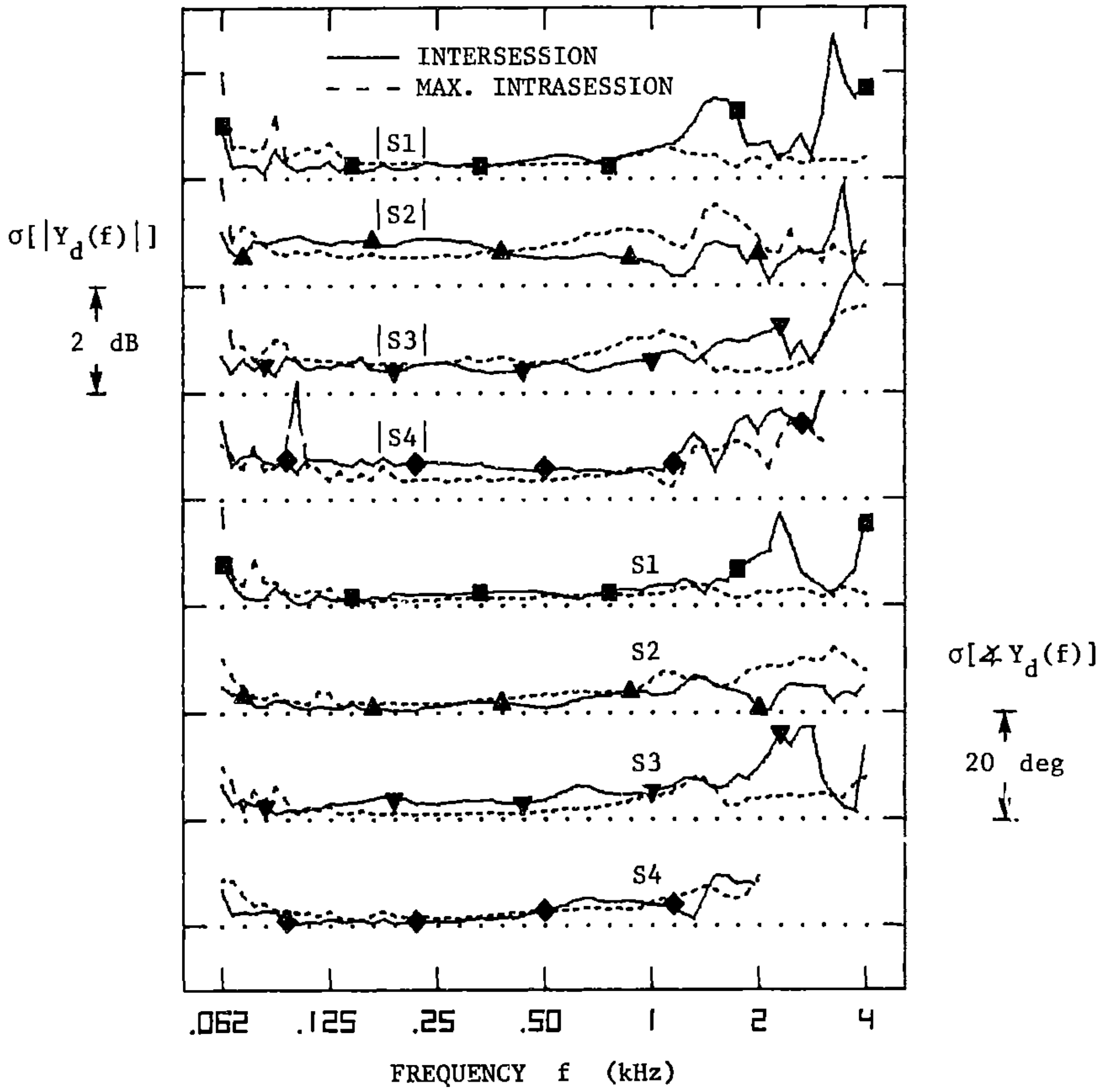

INTERSESSION
MAX. INTRASESSION
|S1|
|S2|
|S3|
|S4|
S1
S2
S3
S4
σ[|Y_d(f)|]
2 dB
σ[∡Y_d(f)]
20 deg
.062
.125
.25
.50
1
2
4
FREQUENCY f (kHz)

FIGURE 18

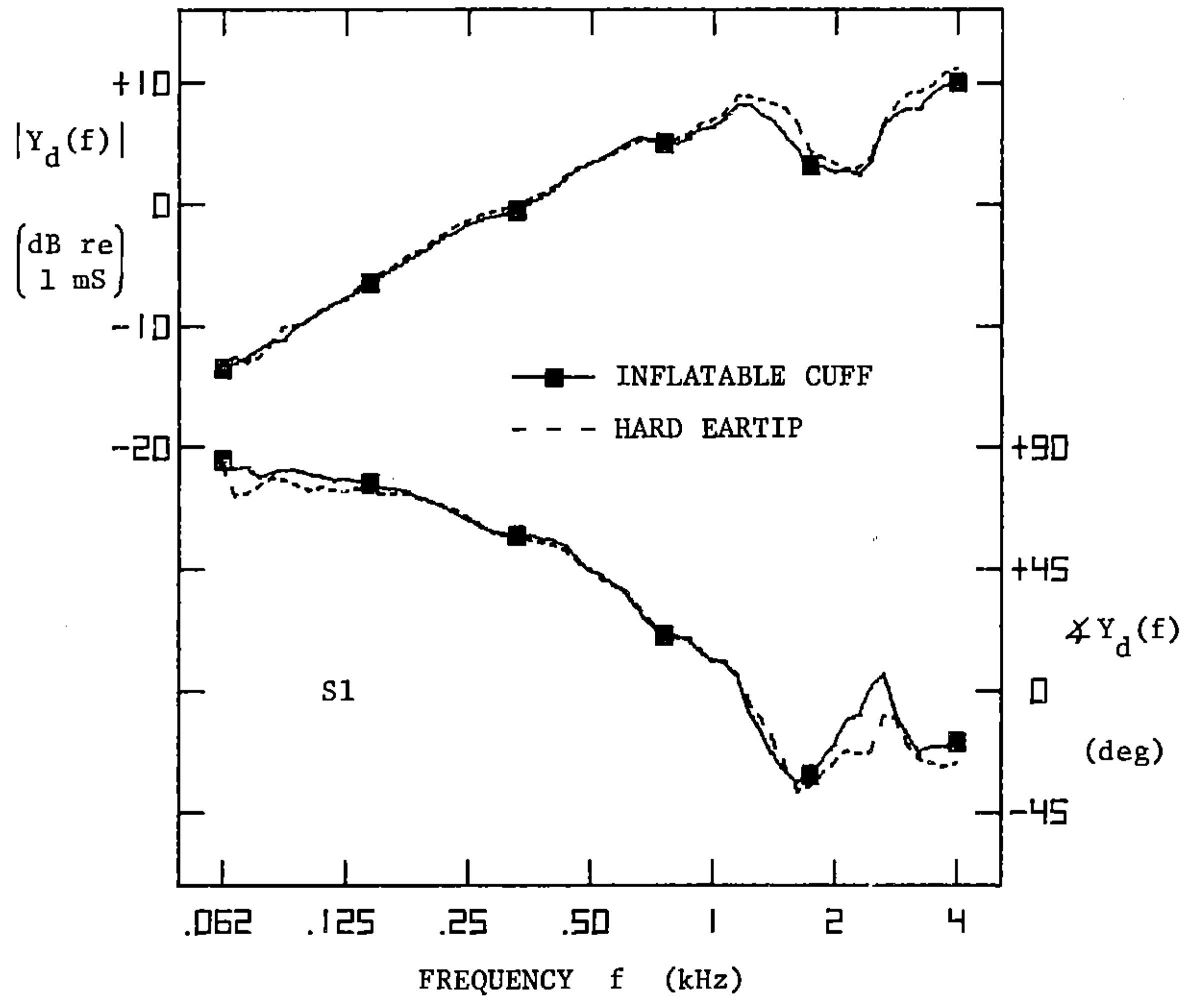
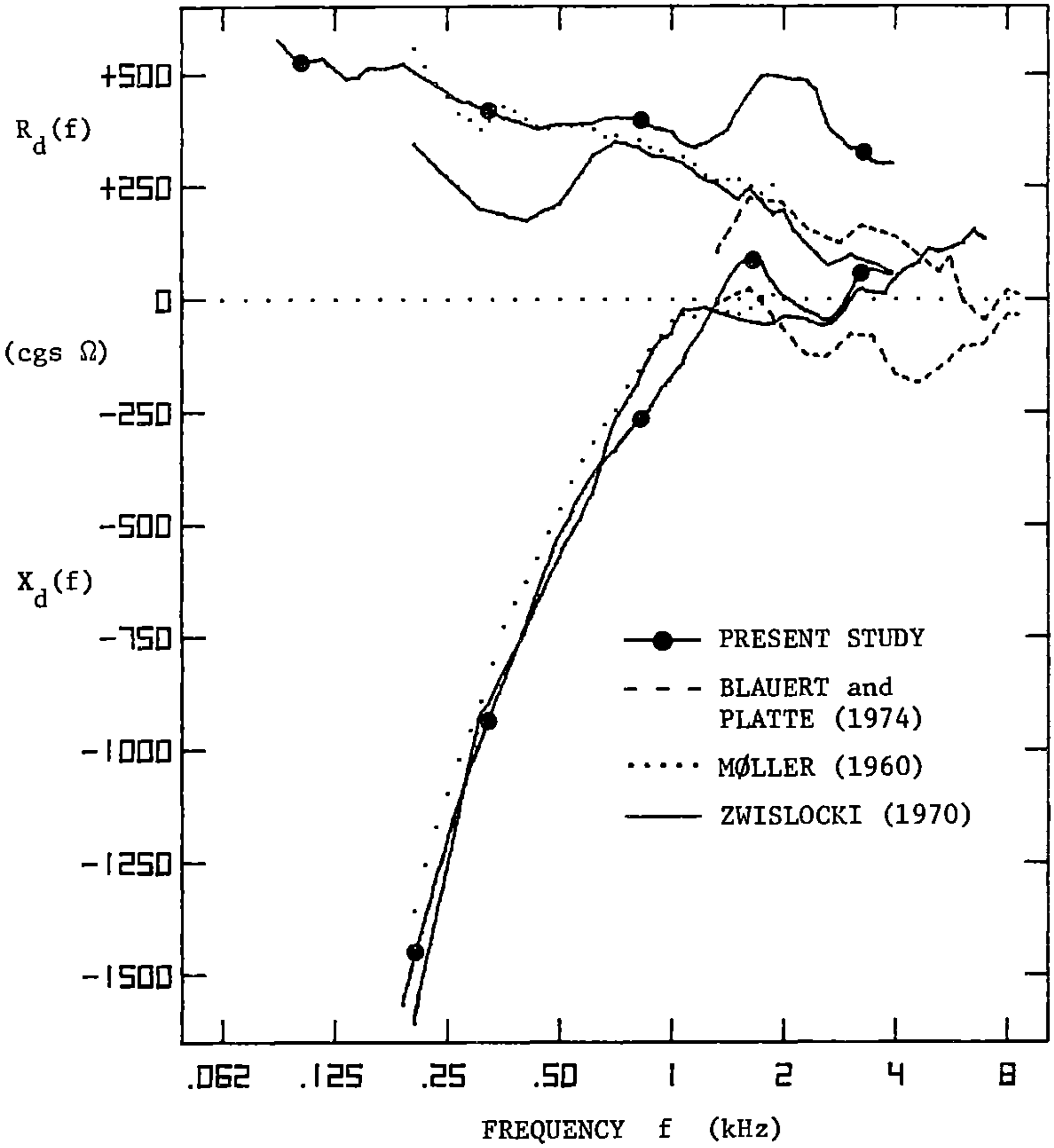

+10
|$Y_d$(f)|
0
(dB re 1 mS)
-10
-20
INFLATABLE CUFF
HARD EARTIP
+90
+45
∡$Y_d$(f)
0
(deg)
-45
S1
.062
.125
.25
.50
1
2
4
FREQUENCY f (kHz)

FIGURE 19



+500
$R_d(f)$
+250
0
(cgs Ω)
-250
-500
$X_d(f)$
-750
-1000
-1250
-1500
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)
PRESENT STUDY
BLAUERT and PLATTE (1974)
MØLLER (1960)
ZWISLOCKI (1970)

FIGURE 20 (a)

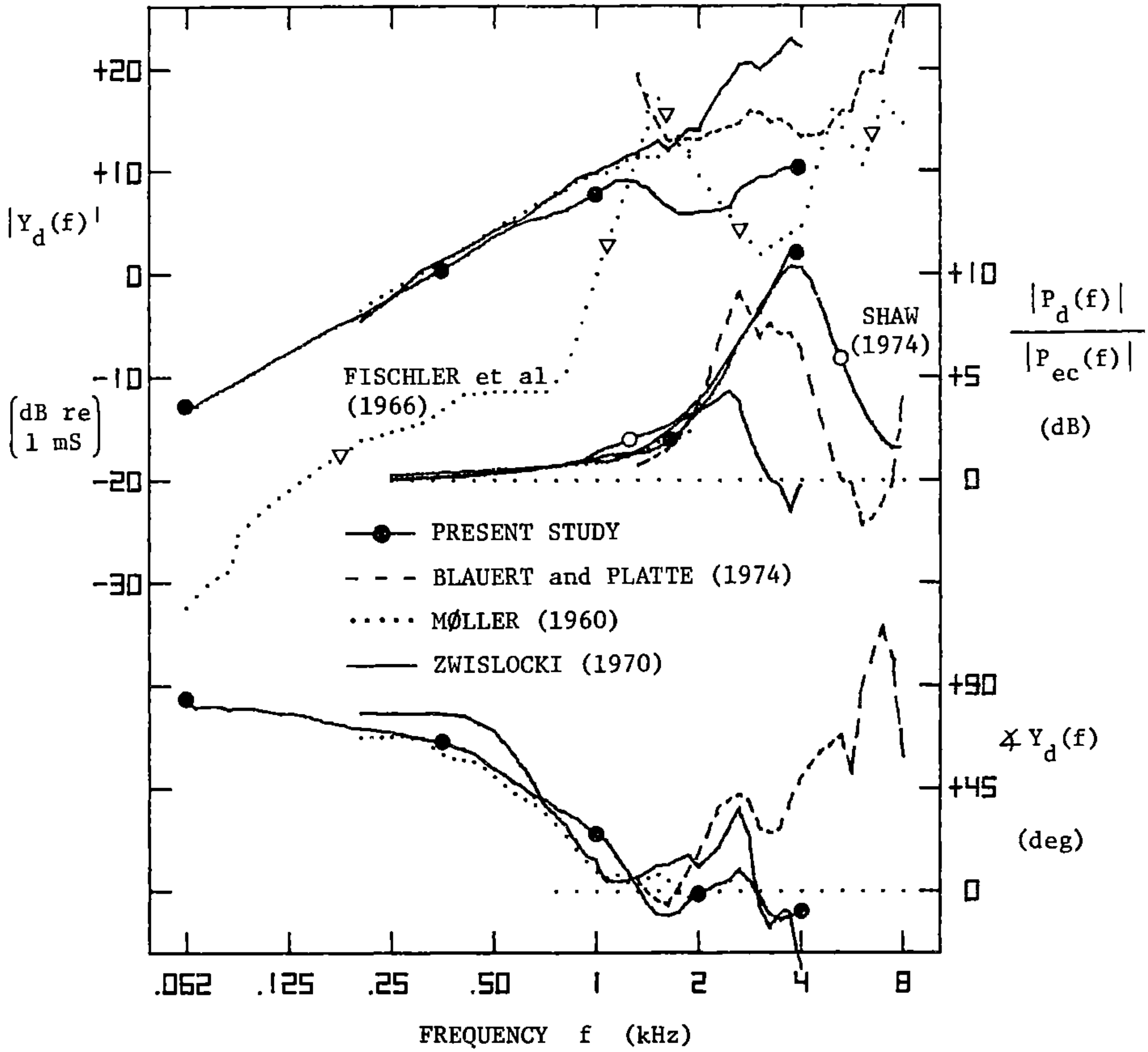
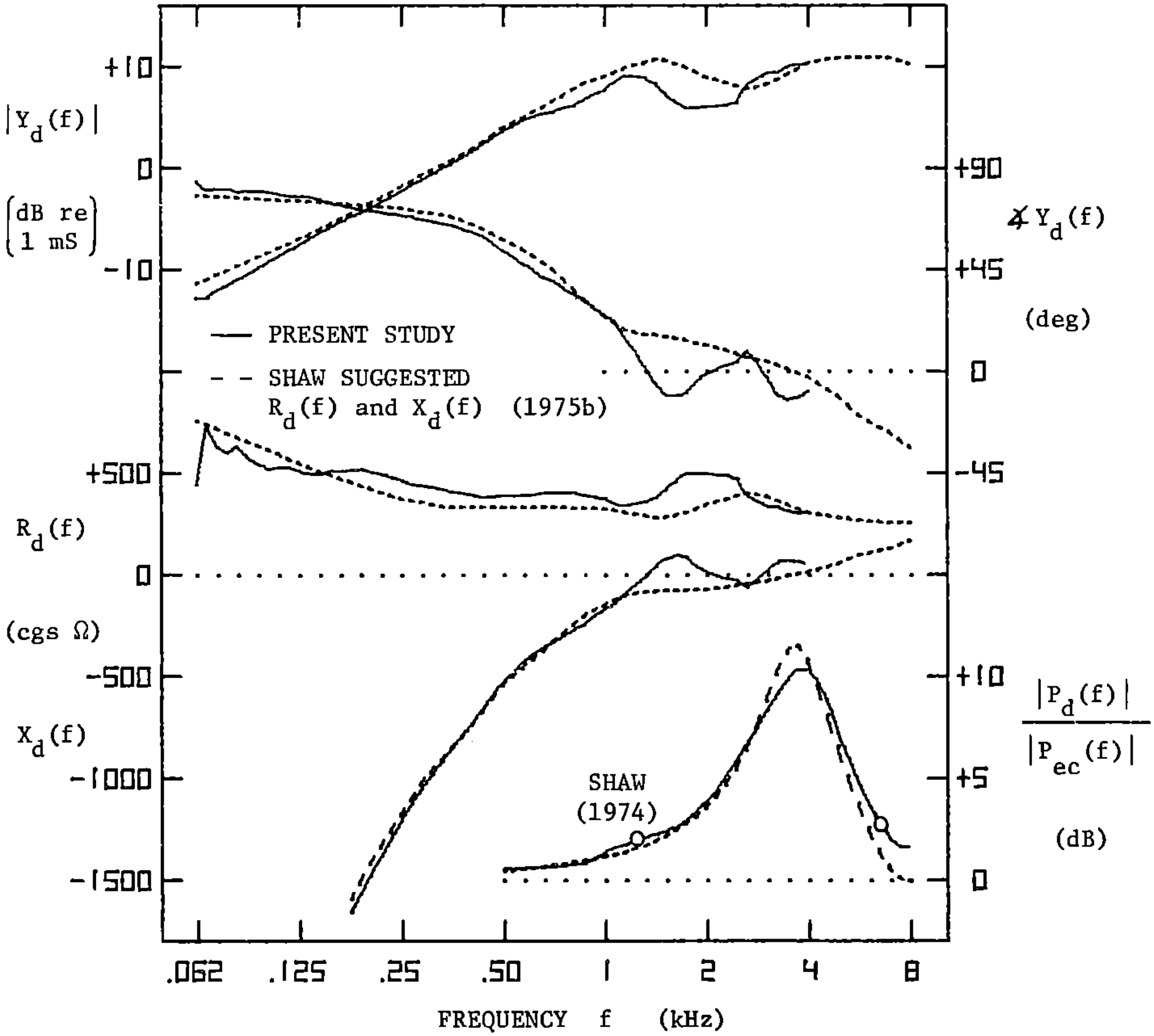
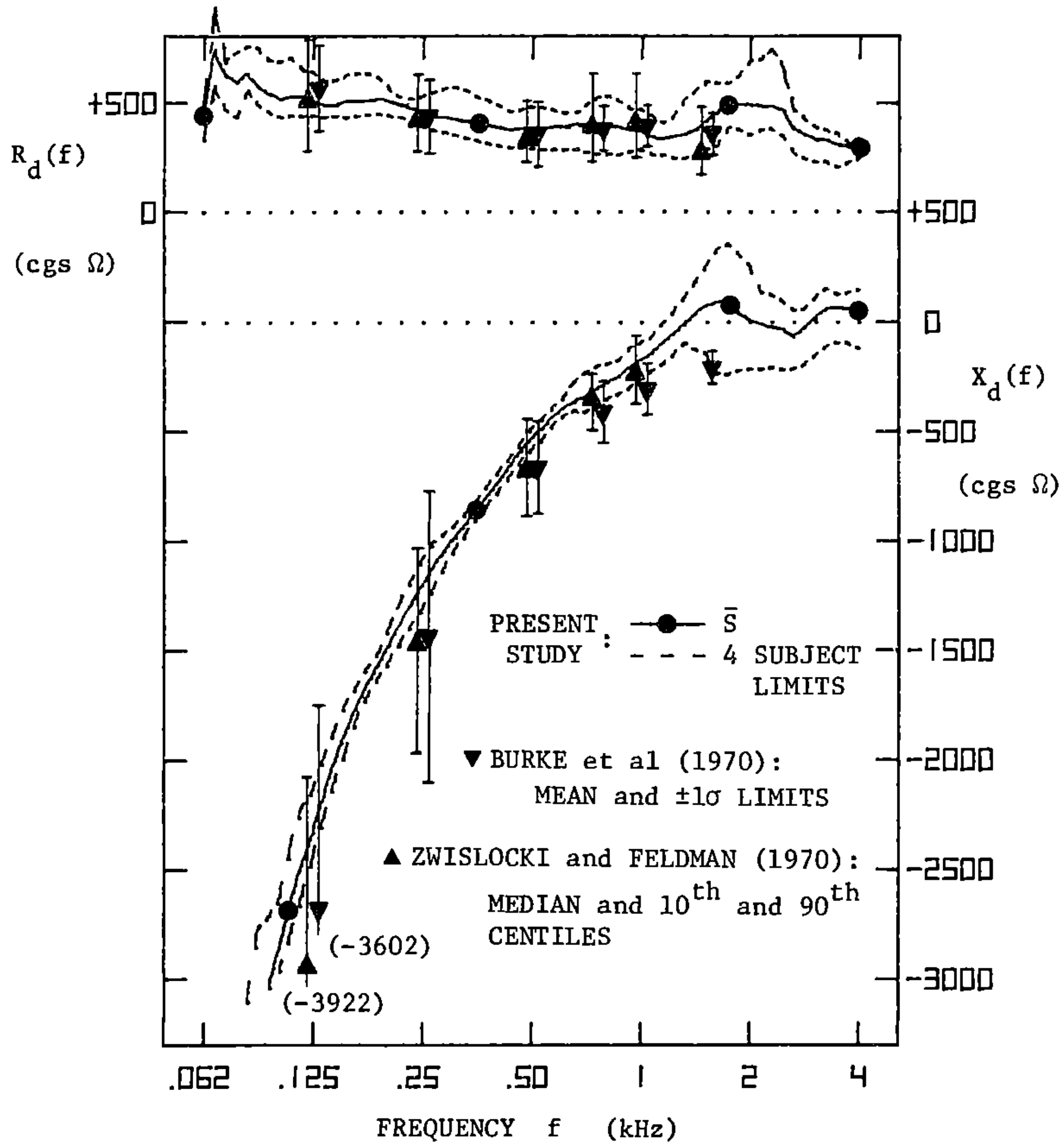

+20
+10
$|Y_d(f)|$
0
-10
(dB re 1 mS)
-20
-30
+10
+5
0
+90
+45
0
SHAW (1974)
$\frac{|P_d(f)|}{|P_{ec}(f)|}$
(dB)
FISCHLER et al (1966)
PRESENT STUDY
BLAUERT and PLATTE (1974)
MØLLER (1960)
ZWISLOCKI (1970)
$\measuredangle Y_d(f)$
(deg)
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 20(b)



+10
|Y_d(f)|
0
(dB re 1 mS)
-10
+90
∡Y_d(f)
+45
(deg)
— PRESENT STUDY
- - SHAW SUGGESTED R_d(f) and X_d(f) (1975b)
0
+500
-45
R_d(f)
0
(cgs Ω)
-500
+10
|P_d(f)| / |P_ec(f)|
X_d(f)
-1000
+5
SHAW (1974)
(dB)
-1500
0
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 21



+500
R_d(f)
0
(cgs Ω)
+500
0
X_d(f)
-500
(cgs Ω)
-1000
-1500
-2000
-2500
-3000
PRESENT STUDY : —●— S̄
- - - 4 SUBJECT LIMITS
▼ BURKE et al (1970): MEAN and ±1σ LIMITS
▲ ZWISLOCKI and FELDMAN (1970): MEDIAN and 10th and 90th CENTILES
(-3602)
(-3922)
.062
.125
.25
.50
1
2
4
FREQUENCY f (kHz)

FIGURE 22

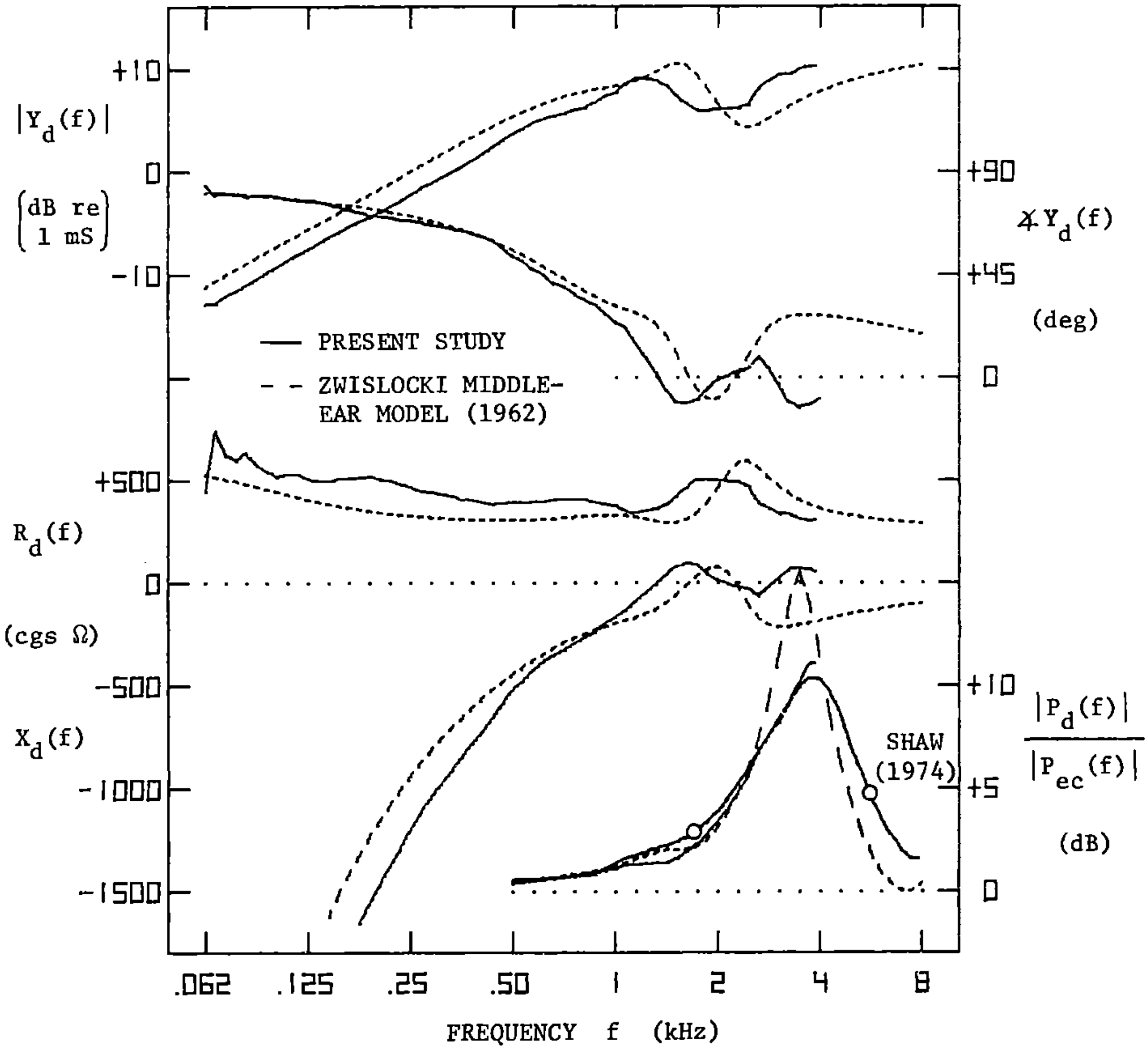

+10
|Y_d(f)|
0
(dB re 1 mS)
-10
+90
∡Y_d(f)
+45
(deg)
0
PRESENT STUDY
ZWISLOCKI MIDDLE-EAR MODEL (1962)
+500
R_d(f)
0
(cgs Ω)
-500
X_d(f)
-1000
-1500
+10
|P_d(f)| / |P_ec(f)|
SHAW (1974)
+5
(dB)
0
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 23

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# PART 2.

# ACOUSTIC-REFLEX EFFECTS ON THE INPUT ADMITTANCE AND TRANSFER CHARACTERISTICS OF THE HUMAN MIDDLE-EAR

## I. Introduction

The occurrence of contractions of the middle-ear muscles (MEM), and particularly the stapedius, in response to high intensity sound is well established. On animals, measurements have been made for the effects of these contractions on both the input admittance (or impedance) and the transmission properties of the middle-ear (Møller, 1965). In humans, extensive measurements have been made of changes in input admittance resulting from acoustically-elicited MEM contractions (i.e. "acoustic reflexes"), and such measurements are increasingly being used clinically for various diagnostic purposes (c.f. Sec. IV in Northern, 1976). However, because of the difficulty in obtaining a direct measure of middle-ear output (without surgical intervention), measurements of changes in transmission accompanying acoustic reflexes in humans have been indirect and inconsistent (c.f., Ward, 1974; Morgan and Dirks, 1975); thus, relations between input admittance- and transmission-changes in humans have not been explored in any detail.

The present study constitutes an attempt to measure and relate both of these quantities in human subjects.

A psychophysical method of measuring (middle-ear) transmission-change has been developed using the sensitivity of the phase of the $2f_1-f_2$ aural combination tone to stimulus level. The method provides frequency-specific estimates of the magnitude of transmission-change. Results have been obtained for frequencies from 400 Hz to 4 kHz with acoustic reflexes elicited by contralateral band-pass noise at levels up to 110 dB SPL.

Physical measurements of absolute admittance-change at the eardrum (62 Hz to 4 kHz) have been obtained for the same subjects from admittance measurements with and without acoustic reflexes.

## II. The reflex-eliciting stimulus

### II. A. Selection of 2 to 4 kHz band-pass noise

In selecting 2 to 4 kHz band-pass noise as the reflex-eliciting stimulus, two factors were considered: (1) Noise was chosen instead of a single tone because noise results in a lower stimulus level being required to elicit acoustic reflexes (see review in Sec. III. C). Low eliciting-stimulus levels were desirable to permit the study of supra-threshold reflex effects without requiring levels (a) that would be difficult to deliver to the subjects' ears, and (b) that would be unpleasant to the subjects. (2) A reflex-eliciting stimulus composed of high frequencies was selected because of a property of the procedure used to measure transmission-changes (see Sec. IV). Specifically, perception of the $2f_1-f_2$ aural combination tone generally becomes more difficult at low frequencies (i.e., for values of $f_1$ below 500 Hz). Hence, an elicitor composed of high frequencies was selected in an attempt to reduce the effects of contralateral masking (associated with the intense elicitor levels) on the transmission-change measurements at low frequencies.

### II. B. Earphones, housings, and calibration

Two different earphones and housings were used for delivering the reflex-eliciting stimulus to the (left) ear. For the admittance-change studies, the elicitor was delivered through a Telephonics

TDH-39 (300 Ω) earphone mounted within the circumaural earcup of a Mine Safety Appliances Noisefoe Mark IIM aural protector. This protector was one side of the headset which supported our admittance measurement system in the right ear (see Sec. II. B of Paper 1). For the transmission-change studies, the elicitor was delivered through a TDH-39 (300 Ω) earphone mounted in a Grason-Stadler GS-001A circumaural cushion (and a Telephonics TC-89E headband).

A relative calibration between these two systems was required so that the results of the admittance- and transmission-change studies could be compared with respect to elicitor level: Each earphone/housing was placed on an Extended NBS-9A coupler (Charan et al, 1965). With the 2 to 4 kHz noise-elicitor, the two systems were equivalent. Specifically, for equal input voltages, both systems developed equal (within 0.5 dB) sound-pressure levels and were relatively insensitive (±1.5 dB) to displacements of 1 cm re center placement on the coupler plate and/or to changes in coupling force caused by adding 0 or 800 g to the self-weight of the earphone/housing systems (mass of TDH-39/GS-001A = 210 g, mass of TDH-31/MSA earcup = 245 g).

An absolute calibration was specified based on testing the TDH-39 earphone by itself, on standard 6 cc couplers.[1] An input of 1 volt rms of the 2 to 4 kHz noise-elicitor developed 109 dB SPL on an ASA Type I coupler and 111 dB SPL on an NBS-9A coupler. As a result, a nominal absolute sensitivity of 110 dB SPL for 1 volt input was adopted.[2]

## III. Admittance-change measurements

### III. A. Methods

Detailed descriptions of the instrumentation, experimental procedures, and computational procedures used to obtain estimates of the "resting" admittance at the eardrum from 62 Hz to 4 kHz are given in Part 1. To obtain admittance measurements over the same frequency range while eliciting acoustic reflexes, those methods were modified as follows:

The computer-controlled frequency-response system (see Sec. II. B. of Part 1) which controlled the "sweeps" of (admittance-measuring) tone-pulses was modified to also provide for the delivery of (reflex-eliciting) noise-pulses to the contralateral ear. The (2 to 4 kHz) noise was gated in time and adjusted in level using an additional electronic switch and programmable attenuator.

For these "reflex-sweeps" the timing sequence shown in Fig. 3 of Part 1 was modified as follows: The noise was turned on (20 msec rise-time), 50 msec elapsed, and then the tone was turned on. As for the sweeps without noise, the admittance measurement (a) did not start until 330 (or sometimes 530) msec following the tone onset, and (b) was made over an interval of 250 msec. Relative to the noise onset, there fore, the admittance measurement began 380 (or 580) msec later. Following the admittance measurement, the noise and tone were simultaneously turned off (20 msec fall-times). The duration of each noise pulse was 650 (or 850) msec. The interval between successive noise pulses was 600 msec.

The level of the noise pulses was constant for each "reflex-sweep"; maximum levels of 110 and sometimes 115 dB SPL were tested.

The four normal-hearing subjects who served for the "resting" admittance measurements of Part 1 of this thesis also served for the present measurements. Furthermore, all of the admittance measurements reported herein were obtained during the sessions reported on in Part 1. These same subjects served for the transmission-change measurements discussed in Sec. IV.

## III. B. Results

Fig. 1 shows typical results for the four subjects of the dependence of admittance-change on frequency and on the level of the reflex-elicitor (denoted $L_N$). The standard deviations of the resting admittances, $\sigma[Y_d(f)]$, are also shown.

Although substantial differences exist in the values of the admittance-changes for the different subjects, the form of the changes is quite uniform. For the more intense elicitors the pattern of the magnitude-changes consists of (1) a relatively constant reduction at low frequencies, (2) a sharp increase in the vicinity of 0.5 to 1 kHz, and (3) positive magnitude-change at high frequencies. The angle-changes are basically positive with a maximum in the frequency region where the magnitude undergoes a sharp upward transition. Below 250 Hz and above 2 kHz the angle-changes are near zero. The sizes of the maximum angle-changes and the low-frequency magnitude reductions are similarly ordered across the four subjects with $S1 > S2 > S3 \simeq S4$. The standard deviations of the resting admittance estimates are relatively small compared to the size of the admittance-changes for the maximum $L_N$'s. These standard deviations give a general indication as to the significance of the admittance-change estimates and also show that no long-term admittance-changes resulted from the "reflex" sweeps.

The dependence of the admittance-changes on $L_N$ is similar for all subjects: First, there is a threshold value of $L_N$ for which admittance-change is just detectable. The data in Fig. 1 indicate a threshold value of 85 to 90 dB SPL. (95 dB SPL is above threshold as each subject shows clear admittance-change for this $L_N$.) Second, a tendency for saturation of admittance-change is evident. Specifically, for $L_N \geqslant 105$ dB SPL (i.e. the data marked with filled symbols) the admittance-changes, in magnitude and angle, nearly superimpose. We shall refer to this admittance-change as the "maximum admittance-change", MAX-$\Delta Y_d(f)$. [The saturation appears most completely and clearly for S1 and S2; for S3 some growth in the admittance-change occurs as $L_N$ increases from 105 to 110 dB SPL.] Each of the curves in Fig. 1 was derived from a single sweep for each value of $L_N$. Similar measurements on different days demonstrated the repeatability of both the reflex threshold values and the near-saturation of admittance-change. The intersubject variations in the degree of $\Delta Y_d(f)$-saturation were also repeatable.

Results for the MAX-$\Delta Y_d(f)$ for the four subjects are compared in Fig. 2. Because of the small size of the standard deviations and because these results are for conditions in which the admittance-change is insensitive to changes in $L_N$, we conclude that the intersubject differences are real. Three features of these differences are: (1) The frequency region in which the transition to positive magnitude-change [and the maximum angle-change] occurs varies from about 0.5 kHz (for S3) to 1 kHz (for S4). (2) The size of the low-frequency magnitude reductions range from about 1 dB (for S3 and S4) to 5 dB (for S1). (3) The maximum angle-changes vary from about +20 degrees (for S3 and S4) to 40 degrees (for S1). As noted earlier, the latter two features are similarly ordered across the four subjects; however, these two features appear uncorrelated with the transition frequencies (e.g. compare S3 and S4).

From these results an average MAX-$\Delta Y_d(f)$ was calculated (the heavy line for $\bar{S}$ in Fig. 2). In Fig. 3, the relationship of this average <u>change</u> to the <u>total</u> admittance at the eardrum is shown; specifically, we plot the total admittance with and without the reflex active.

The preceeding plots have represented the admittance as a complex quantity in terms of its magnitude and phase angle. Alternate forms of the data might provide additional insight into the reflex effects. The data from Fig. 3 are displayed (a) in Fig. 4, as the conductance, $G_d(f)$, and susceptance, $B_d(f)$, which are the real and imaginary parts of the admittance $Y_d(f) = G_d(f) + jB_d(f)$, and (b) in Fig. 5 as the resistance, $R_d(f)$, and reactance, $X_d(f)$, which are the real and imaginary parts of the impedance $Z_d(f) = R_d(f) + jX_d(f) = 1/Y_d(f)$. Fig. 5 makes it clear that the reflex produces a relatively large change in the reactance, $X_d(f)$, below 2 kHz, and small frequency-dependent changes in the resistance, $R_d(f)$.

In Fig. 6, the impedance-changes, $\Delta R_d(f)$ and $\Delta X_d(f)$[3], are plotted for the individual subjects and for $\bar{S}$. The resistance-change, $\Delta R_d(f)$, is small for each of the four subjects, but it is systematic in that it is positive at low-frequencies and it decreases gradually with increasing frequency. For each subject, $\Delta R_d(f)$ does become negative, but the frequency for which the transition from positive to negative $\Delta R_d(f)$ occurs varies from about 250 Hz (for S4) to about 1 kHz (for S3).

## III. C. Discussion

### (1) Reflex threshold and dynamic range

As a function of elicitor level, $L_N$, the admittance-change exhibited a limited dynamic range of about 15 to 20 dB [i.e., no admittance-change was detectable for $L_N$ below 85 to 90 dB SPL, and nearly-constant admittance-change (at a given measurement frequency) occurred for $L_N$ above 105 dB SPL]. That a limited dynamic range exists is well known (c.f., Møller, 1961b, 1962a, 1962b; Dallos, 1964, 1973), and our values for the dynamic range and reflex threshold can be compared to those reported by others.

The reflex threshold depends upon the spectral composition and duration of the eliciting stimulus, and on the time after elicitor onset that the admittance measurement is made. For single-tone elicitors (of durations greater than ~0.5 sec) the reflex threshold is approximately 75 to 90 dB above the threshold of audibility (c.f., Møller, 1962a; Jepsen, 1963; Franzen, 1970). Increases in the bandwidth of an eliciting stimulus result in decreases in the rms level of the stimulus at the reflex threshold. These decreases in threshold are small (-3 to -6 dB) up to the "critical bandwidth" for the acoustic reflex, and then the threshold decreases more rapidly (-3 to -12 dB per octave) for further increases in elicitor bandwidth (Møller, 1962b; Flottorp et al, 1971; Djupesland and Zwislocki, 1973; Popelka et al, 1974; Popelka et al, 1976). In comparison to critical bandwidths measured psychophysically (c.f. Scharf, 1970), the critical bandwidths for the acoustic reflex are larger, being about 0.7 to 1.0 times the center frequency of the eliciting stimulus.

Our paradigm did not allow a very precise determination of the reflex threshold, however, our results do indicate a threshold value of 85 to 90 dB SPL. Results in the literature (Flottorp et al, 1971; Peterson and Lidén, 1972; Margolis and Fox, 1976; Popelka et al, 1976) for elicitors that are spectrally comparable to our 2 to 4 kHz noise indicate average thresholds in the range of 79 to 86 dB SPL (average across studies $\approx$ 83 dB SPL).[4] These values are close to ours, but they are suggestive that our threshold value is a few dB above that which might be expected solely on consideration of the spectral composition of the eliciting stimuli. This discrepancy can roughly be accounted for by consideration of the timing parameters used, and the criteria used for defining the reflex threshold. Specifically, our elicitor was 650 msec in duration, and in making the admittance measurements the microphone output was averaged over a 250-msec interval beginning 380 msec following the elicitor onset (see Sec. III. A). In t four studies cited above, longer duration elicitors were used (2.4, 2.0, 1.0, and 2.0 sec respectively), and the reflex threshold was defined as the elicitor level resulting in any admittance-change time-locked to the elicitor presentation. The onset response of admittance-change is known to progressively decrease in speed with decreases in elicitor level, and for near-threshold elicitors (a) the admittance-change is often irregular (not constant with time) and (b) the maximum admittance-change may not occur within 1 sec after elicitor onset (Møller, 1962b; Dallos, 1964, 1973; Lilly, 1964; Hung and Dallos, 1972). Additionally, studies on temporal summation for the reflex threshold indicate that lower thresholds occur for elicitor durations at least up to 1 sec (Djupesland and Zwislocki, 1971; Djupesland et al, 1973; Richards, 1975). Hence, because of the eliciting-stimulus duration used in our tests and the fact that our admittance measurements were averaged over a fixed and relatively long time interval, we expect that (as observed) our threshold values should be above those obtained in studies cited above.

In comparison to the number of studies on reflex thresholds, measurements of the growth of admittance-change vs elicitor level are more limited. In particular, we are unaware of reports of "reflex growth" with eliciting stimuli composed of high-frequency bands of noise. However, some measurements by Møller (1962b, Figs. 6 and 7) with elicitors at lower frequencies (525 and 1450 Hz) suggest that roughly similar reflex-growth occurs for single-tone and band-pass elicitors.

Our measurements indicate reflex-growth occurs over an elicitor-level range of 15 to 20 dB. For single-tone elicitors in the range of 2 to 4 kHz, measurements and discussion in Møller (1962a, 1962b), Djupesland et al (1967), Peterson and Lidén (1972), and Beedle and Harford (1973) suggest a dynamic range averaging about 20 dB. Because of the timing parameters used in our tests (as discussed above), it was suggested that our estimate of the reflex threshold was "biased" high by a few dB. For the two reasons which follow, we suggest that the elicitor level measured for near-saturation of admittance-change, $L_N$ = 105 dB SPL, should not have been "biased" in our tests; hence, our estimates of the dynamic range could be expected (as observed) to be low by a few dB. (1) For elicitors above ~15 dB re reflex threshold, the "steady-state" admittance-change should be reached within the 380 msec we allowed after elicitor onset and before admittance measurement (see onset response measurements in Møller, 1962b; Dallos, 1964, 1973; Lilly, 1964; Hung and Dallos, 1972). (2) For the combination of high elicitor levels and successive elicitor pulses separated by 0.6 sec (as used in our sweep measurements), MEM contraction relaxes only partially between the successive elicitor pulses [as inferred from results (Hung and Dallos, 1972) obtained with sinusoidally-modulated elicitors]. Hence, the "steady-state" admittance-change for successive elicitor pulses is probably reached (within each pulse) more rapidly than that for an isolated (650-msec) elicitor pulse.

In summary, our values of $L_N$ = 85 to 90 dB SPL for the (reflex) threshold of admittance-change and $L_N$ = 105 dB SPL for the near-saturation of admittance-change are reconcilable with measurements in the literature; in Sec. V these values are used in relating admittance-change vs elicitor level and transmission-change vs elicitor level.

### (2) Size of the "maximum admittance-change"

For comparison to our results for the "maximum admittance-change", MAX-$\Delta Y_d$(f), we have considered data on absolute admittance-change (or impedance-change) reported in Møller (1961a), Dallos (1964), Lilly and Shepherd (1964, as given in Feldman and Zwislocki, 1965), Feldman and Zwislocki (1965), Terkildsen et al (1970), and Hung and Dallos (1972). From these studies eliciting stimuli of various frequency compositions and levels were employed; for comparison with our MAX-$\Delta Y_d$(f) we have used the largest reported changes. In all of these studies the frequency range over which the admittance measurements were made is more limited than ours. The comparisons are made separately for frequencies below 500 Hz, for frequencies between 500 and 2000 Hz, and for frequencies above 2 kHz.

At frequencies below 500 Hz, a reduction of admittance magnitude is consistently reported. However, variation exists in the size of the reductions. Our average reduction of 2.5 dB (with a range of about 1 to 5 dB) is in agreement with mean results (at 250 Hz) of 2.4 dB for Lilly and Shepherd, and 2.7 dB for Feldman and Zwislocki. Somewhat larger mean reductions of 4.0 dB are obtained from the data of Dallos (at 250 Hz, with a range of 2.5 to 6.8 dB) and Terkildsen et al (at 220 Hz, with a range of 1.5 to 6.8 dB), whereas a smaller mean reduction of 1.0 dB (with a range of 0.5 to 2.2 dB) can be calculated from the data of Hung and Dallos (at 400 Hz).

A second feature of our results at low frequencies is that the changes in the resistance, $\Delta R_d(f)$, are small in comparison to the changes in the reactance, $\Delta X_d(f)$, and that below about 250 Hz, $\Delta R_d(f)$ is positive for all four subjects. Other authors have also noted the small size of $\Delta R_d$ re $\Delta X_d$, but the sign of $\Delta R_d$ varies from one report to another. Møller's results indicate that $\Delta R_d(f)$ is positive below ~400 Hz. However, Feldman and Zwislocki (at 125 and 250 Hz) report mean changes that are negative (with $\Delta R_d \approx -50$ ohms) and individual results that are both negative and positive. The results in Dallos (at 250 Hz) also indicate changes that are mixed with $\Delta R_d$ negative for four subjects and positive for two subjects. Lilly and Shepherd report positive mean changes, but their results appear atypical in that the resistance-changes at 125 and 250 Hz are large, $\Delta R_d \approx +400$ ohms, and equal to the size of the reactance-changes.

At frequencies from 500 to 2000 Hz, an important aspect of our results is that similarities occur in the form of the admittance-changes for all of our subjects, but that the changes do not occur at the same frequencies. Our subjects indicated an upward transition to a region of positive magnitude-change and an associated maximum positive angle-change. [In terms of the real and imaginary components of the impedance, the results indicated a maximum negative resistance-change and a negative reactance-change decreasing toward zero with increasing frequency.] The subjects differ by about one octave in the frequencies for which these effects occur. In the data obtained with the Zwislocki acoustic bridge (Dallos; Lilly and Shepherd; Feldman and Zwislocki) some of these same features are evident; however, the consistent form that we observe has not been emphasized because (a) the measurements are generally restricted to an upper frequency limit of 1000 or 1500 Hz (resulting from the inability of the bridge to balance positive reactances), and (b) the measurements have been made at fewer frequencies. Møller's results, extending to 1800 Hz with measurements each 50 Hz, exhibit the same form as our results, but the sizes of the reported changes exceed ours somewhat. For example, near 1300 Hz the increase of admittance magnitude ranges from 2.5 to 4.5 dB in Møller's data whereas our increases are 1 to 2 dB. Similarly, near 1000 Hz the largest resistance reduction in Møller's data, about 300 ohms, is about twice as large as our maximum reductions of 120 to 160 ohms. [Feldman and Zwislocki report mean reductions of 90 ohms at 750 and 1000 Hz.]

At frequencies above 2 kHz, we are unaware of any comparable measurements of admittance-change; our results suggest that the changes are near zero.

In general, the agreement between the results of the various studies and our results for MAX-$\Delta Y_d(f)$ is good. All authors indicate that large intersubject variability exists; this variability may be responsible for some of the differences we have noted between the studies However, the non-overlap in the ranges of some of the individual results [e.g., the low-frequency admittance magnitude-reductions reported by Dallos, 2.5 to 6.8 dB (N = 6), and Hung and Dallos, 0.5 to 2.2 dB (N = 4) suggests that some inaccuracies may be present in the reported data.

These inaccuracies may derive from having biased estimates of the residual earcanal volume between the tip of the measuring device and the eardrum (see Fig. 7), and furthermore in the study by Hung and Dallos, from having used simplifying assumptions (as detailed in Pinto and Dallos, 1968) to compute the impedance-changes at the eardrum.

## IV. Transmission-change measurements

### IV. A. Methods

#### (1) Basic scheme

A psychophysical procedure (Rabinowitz and Goldstein, 1973) which exploits the sensitivity of the phase of the aural combination-tone $2f_1-f_2$ to stimulus level was used to infer the transmission-change produced by acoustic reflexes. Prior to outlining details of the method, we first review relevant properties of the $2f_1-f_2$ aural combination tone.

With acoustic stimuli composed of two tones of frequencies $f_1$ and $f_2$, one readily hears a pitch corresponding to the frequency $2f_1-f_2$ (if the levels and frequencies of the two tones are chosen appropriately). This $2f_1-f_2$ "aural combination tone" has recently been the subject of much research, which was stimulated by the work of Goldstein (1967). Goldstein measured the distortion tone psychophysically, by adding an externally generated tone of frequency $2f_1-f_2$ (to the two primary tones $f_1$ and $f_2$ with levels $L_1$ and $L_2$) which the subjects adjusted in amplitude and phase to "cancel" the subjective $2f_1-f_2$ component. The level $L_c$ of the "cancellation tone" grows almost linearly with primary stimulus level from 20 to 70 dB SL (with $L_1 = L_2$, and $f_2/f_1$ in the range 1.1 to 1.5). Furthermore, $L_c$ is relatively large for $f_2/f_1 = 1.1$ (typically $L_c/L_1 = -14$ dB, i.e. the combination tone is about 20% of the primary tone), but $L_c$

decreases rapidly at about -100 dB/oct as $f_2/f_1$ increases. In subsequent psychophysical studies, these basic properties have been confirmed and some ranges of primary stimulus parameters leading to "irregular" behavior of the cancellation tone have also been identified (c.f., Helle, 1969; Goldstein, 1970; Hall, 1972a, 1972b, 1975; Smoorenburg, 1972a, 1972b; Zwicker and Fastl, 1973; Weber and Mellert, 1975; Buunen et al, 1977). In addition, physiological experiments have searched for correlates of the $2f_1-f_2$ tone in basilar membrane motion (Wilson and Johnstone, 1973), in cochlear microphonic potentials (Dallos, 1969), in the discharge patterns of auditory nerve fibers (Goldstein and Kiang, 1968), in the discharge patterns of cochlear nucleus cells (Greenwood et al, 1976; Buunen et al, 1977), and in theoretical studies of cochlear mechanisms (c.f., Hall, 1974; Schroeder, 1975a, 1975b; Duifhuis, 1976; and for extensive material and review, the section on "Nonlinear Effects" in Zwicker and Terhardt, 1974). Although the generating mechanism of the physiological correlate to the psychophysically measured $2f_1-f_2$ tone is not known, all the work to date suggests that the origin is central to the middle-ear, and probably is within the inner-ear (as specifically discussed in Schroeder, 1975b).

That the $2f_1-f_2$ correlate is generated central to the middle-ear is a requirement of our procedure for deduction of middle-ear transmission-change.

A second property of the combination tone is crucial to our use for measurement of transmission-change: The phase angle of the cancellation tone, $\Theta_c$, strongly depends upon the level $L_1$ of the lower primary stimulus tone. For intermediate levels of $L_1$ (about 40 to 65 dB SL), the cancellation-phase changes 5 to 10 degrees per dB change in $L_1$ (c.f., Goldstein, 1967; Helle, 1969; Smoorenburg, 1972b).

If the combination tone is generated central to the middle-ear, then changes in middle-ear transmission should change the level of the stimulus to the inner ear, and as result, change the phase of the cancellation tone. Hence, we can use the cancellation-phase sensitivity to primary stimulus level for the measurement of middle-ear transmission-change, $\Delta|H(f)|$: The cancellation tone is initially adjusted to null the combination tone and then middle-ear transmission is altered by the acoustic reflex which makes the $2f_1-f_2$ reappear. The $2f_1-f_2$ pitch can be removed once again, while the reflex is "on", with a readjustment of the phase of the cancellation tone along with a small adjustment in cancellation-tone amplitude if necessary.

The cancellation-phase change, $\Delta\Theta_c$, can be predominantly attributed to a change in the magnitude of transmission at the lower primary frequency, i.e. to $\Delta|H(f_1)|$, even though the reflex may affect middle-ear transmission nonuniformly in phase and amplitude at the three frequencies ($f_1$, $f_2$, and $f_c = 2f_1-f_2$) comprising the stimulus complex. With regard to the influence of changes in the phase angles, $\Delta\measuredangle H(f)$, it has been reported that $\Theta_c$ is independent of any linear-plus-constant phase transformation (i.e. $\Delta\measuredangle H(f) = a + b\cdot f$) applied to a stimulus complex (de Boer, 1961; Goldstein, 1965; Buunen et al, 1974). Thus, insofar as the changes in the middle-ear phase function, $\Delta \measuredangle H(f)$, can be approximated by a straight line over the (relatively narrow) frequency range of the stimulus complex, these phase changes have no perceptual effect. Complications in interpretation might also result from changes in the magnitude of middle-ear transmission, $\Delta|H(f)|$, which are not uniform in amplitude across the stimulus complex. However, it has been reported that for a fixed level of $L_1$, $\Theta_c$ is relatively insensitive to changes in $L_2$ (c.f., Goldstein, 1967; Helle, 1969; Smoorenburg, 1972b). We have found that for relatively large values of the primary-tone amplitude ratio ($L_1/L_2 \geqslant 15$ dB), $\Theta_c$ is nearly unchanged for changes in this ratio of ±5 dB. [See Fig. 9 with $f_2/f_1 = 1.22$ for example, and data in Helle (1969, Fig. 7) from which the same conclusion can be reached.] Thus, if $\Delta|H(f_2)| = \Delta|H(f_1)| \pm 5$ dB and $L_2/L_1$ is small enough, the measured $\Delta\Theta_c$ can be attributed solely to $\Delta|H(f_1)|$.

A consistency check of the method was carried out using stimulus complexes with the same $f_1$ but differing values of $f_2/f_1$. Such complexes exhibit different $\Theta_c$ sensitivities to changes in lower primary-tone level, with the slope $\Delta\Theta_c/\Delta L_1$ increasing with the value of $f_2/f_1$ (see Fig. 9). However, equal transmission-changes at $f_1$ (for the same reflex condition) are implied by the measurements with different values of $f_2/f_1$.

Some useful features of the method result from the use of a double-sided, null-detection with the $2f_1-f_2$ pitch sensation returning for phase deviations either side of "cancellation". (1) The effects of observer criterial-bias can be small. Variations in criterion for detecting the combination tone change the size of the (symmetric) phase deviations used in "bracketing" the null position, but do not change the mean value of the null-estimate. (2) The effects of contralateral masking and/or binaural loudness summation involving the contralateral reflex-elicitor (c.f. Morgan and Dirks, 1975) should make the task more difficult, and thereby increase the variability in the adjustment without, however, affecting the mean.

## (2) Apparatus

Fig. 8 shows a block diagram of the equipment used. The combination-tone-complexes (CTC's) were generated from analog equipment including two low-distortion oscillators (Krohn-Hite 4031R) which generated the primary tones $f_1$ and $f_2$. For deriving the $2f_1-f_2$ cancellation tone, precision four-quadrant multipliers (Analog Devices 425K, 0.1% accuracy) and a dual filter set (Krohn-Hite 3343R, -48 dB/oct) were used. A continuously adjustable solid-state phase-shifter (similar to the Grason-Stadler E3520B) and an adjustable attenuator were used for adjusting the cancellation tone.

The CTC's were delivered to the subject's right ear using a headset incorporating Telephonics TDH-39 earphones mounted in Grason-Stadler GS-001A circumaural cushions and a Telephonics TC-89E headband. The CTC earphone was calibrated on an ASA Type I (6 cc) coupler.

### (3) Selection of CTC parameters and experimental protocol

We began by choosing appropriate CTC parameters $f_2/f_1$ and $L_2/L_1$ for each subject and for each test frequency $f_1$ (at one-third octave intervals from 0.4 to 4 kHz). Compromises were involved in the choice of both $f_2/f_1$ and $L_2/L_1$. A low value of $f_2/f_1$ increases the level $L_c$ of the combination tone. Also, a low value of $f_2/f_1$ keeps the frequency range of the CTC small and thereby reduces the method's vulnerability to the influence of nonuniform transmission-change. However, a low value of $f_2/f_1$ yields poorer measurement sensitivity because the slope $\Delta\Theta_c/\Delta L_1$ decreases as $f_2/f_1$ decreases (see Fig. 9). The values of $f_2/f_1$ used were greatest at low frequencies and smallest at high frequencies (with the maximum values for each subject being 1.25 to 1.30 at $f_1 = 400$ Hz and the minimum values being 1.15 to 1.175 at $f_1 = 4$ kHz). For all values of $f_2/f_1$, $L_2/L_1$ was chosen such that the sensitivity of $\Theta_c$ to changes in $L_2/L_1$ was small. Also, $L_2/L_1$ was constrained to be above -25 dB to avoid greatly weakening perception of the $2f_1-f_2$ aural combination tone; i.e., with decreases in $L_2/L_1$, $L_c$ decreases and as a result the $\Theta_c$ nulling accuracy decreases. The values of $L_2/L_1$ which were chosen ranged from -10 to -20 dB. The resulting $\Theta_c$ sensitivity to $\Delta(L_2/L_1)$ in comparison to that associated with $\Delta L_1$ (keeping $L_2/L_1$ constant) was indeed small as required for attribution of $\Delta\Theta_c$ to $\Delta|H(f_1)|$. Specifically, the ratio $\Delta\Theta_c[\Delta(L_2/L_1) = \pm 5 \text{ dB}]/\Delta\Theta_c[\Delta L_1 = \pm 5 \text{ dB}]$ was typically less than 1/6 (always $\leqslant 1/4$ and often zero, see Fig. 9).

Following selection of $f_2/f_1$ and $L_2/L_1$ and the measurements of the "calibration" function $\Theta_c$ vs $L_1$, the effects of the acoustic reflex were examined. The experimenter set $L_1$ and the level of the reflex-elicitor, $80 \leq L_N \leq 110$ dB SPL, and then instructed the subject to begin a "trial". The subject adjusted the phase shifter for $2f_1-f_2$ cancellation, then turned the reflex-elicitor on, adjusted to the new $\Theta_c$ (and sometimes readjusted $L_c$ slightly), and immediately turned the elicitor off so as to minimize the noise exposure and discomfort. The adjustment times for $\Theta_c$ measurement in the presence of the elicitor varied from 5 to 15 sec (following the elicitor onset). In general, the longer adjustment times (i.e., the more difficult judgments) occurred when the transmission-changes were large, namely, for the maximum $L_N$'s and low $f_1$'s. The measurements were made with moderate values of $L_1$, 65 to 75 dB SPL, but for some of the more difficult judgments the experimenter increased $L_1$ to 80 dB SPL in order to facilitate the $\Theta_c$ judgment. The settings made by the experimenter for $L_1$ and $L_N$ (and $L_2/L_1$) were kept from the subjects' view (except for S1, this subject being the experimenter).

### (4) Stability of $\Theta_c$ measurements without contralateral noise

For each test session numerous measurements of $\Theta_c$ were obtained with no reflex activation. In determining the function $\Theta_c$ vs $L_1$ at the start of each session, repetitions of the same $L_1$'s yielded $\Theta_c$ measurements which were typically within 5 degrees of each other. This stability of $\Theta_c$ (for the no-reflex trials) was often maintained throughout the reflex-testing in the remainder of the test session. Although some remarks in the literature are suggestive that $\Theta_c$ measurements can be made repeatably, quantitative reports of $\Theta_c$ stability are sparse. For repeated $\Theta_c$ measurements Hall (1972a, pg. 1866) reports a 95% confidence interval of ±70 degrees, Smoorenburg (1972b, pg. 621) reports that the standard deviation is "smaller than . . . $0.1\pi$", and Buunen et al (1974, Fig. 5) report differences which vary with stimulus conditions and range from near zero to ~20 degrees. These measures of variability are generally larger than ours; the difference can probably be attributed to the smaller range of CTC parameters used in our tests. Specifically, we always used CTC parameters for which the audibility of the $2f_1-f_2$ combination tone was good, i.e. we avoided large values of $f_2/f_1$ and/or low levels of $L_1$. In contrast, Hall, Smoorenburg, and Buunen et al made measurements for some conditions where the $2f_1-f_2$ combination tone approached inaudibility.

Over the duration of some test sessions, somewhat larger changes occurred for $\Theta_c$, up to 10 degrees (and in one case 15 degrees). When changes of this magnitude occurred, $\Theta_c$ measurements at a number of $L_1$ values were then taken, and these measurements revealed that the changes occurred for all values of $L_1$, i.e. the entire function of $\Theta_c$ vs $L_1$ had been shifted. In these cases, the intervening reflex trials were discarded as it was unclear whether the "before" or "after" values for $\Theta_c$ vs $L_1$ were appropriate for transforming the $\Theta_c$ measurements on the reflex trials to estimates of transmission-change. While it is conceivable that a portion of these $\Theta_c$-changes may have derived from "permanent" changes in middle-ear transmission (perhaps resulting from the MEM contractions), no such effects were identified in our admittance-change tests (see Sec. III. B). Furthermore, some of the changes are likely to have resulted simply from variations in the acoustic coupling of the earphone to the ear. These variations arise from shifts of the earphone/cushion position due to head- and facial-musculature-movements (from talking, sneezing, or yawning).

## IV. B. Results

Illustrative data for the inferred effects of the acoustic-reflex on middle-ear transmission are shown in Fig. 10. Data are presented for each subject at one (low) frequency $f_1$. All of these measurements were inferred from the $2f_1-f_2$ cancellation-phase method; a single CTC was used for S2 and S3, two CTC's for S4, and three CTC's for S1. [For the three CTC's used with S1, the cancellation-phase "calibration" functions of $\Theta_c$ vs $L_1$ are shown in Fig. 9, with $L_2/L_1$ = -15 dB.] In computing the transmission-changes shown, the measured $\Theta_c$'s were attributed entirely to $\Delta|H(f_1)|$.

The general course of these results is clear; with increasing elicitor level, $L_N$, increasing transmission-reductions (i.e. attenuations) occur. Furthermore, the transmission-change data obtained with the different CTC's are in reasonable agreement. The differences with the different CTC's are generally no larger than the differences from repetitions within a single CTC (e.g., see S1 and S4 in Fig. 10). This consistency supports the validity of the psychophysical method, and in the following treatment of the data, measurements made with different CTC's (with the same $f_1$) will be lumped together.

The variability in the data may derive from at least three sources: (a) intrinsic variability in transmission-change for a given value of $L_N$, (b) variability in transmission-change associated with the variable time of combination-tone cancellation relative to the onset of the reflex-elicitor, and (c) variability resulting from the subjects' lack of precision in setting the cancellation-phase. Of these factors we suggest that (c) is certainly significant, (b) is of lesser significance, and (a) is of unknown significance.

The significance of (c) for the high $L_N$ conditions is evident both from the reports of the subjects indicating difficulty in detecting the cancellation-nulls, and (objectively) from the increased range of phase-shifter excursions required in bracketing the null-estimates. Furthermore, in some cases the task was so difficult that no data could be obtained at 110 dB SPL. In general, the trials with $L_N$ = 110 dB SPL were judged to be rather unpleasant by the subjects, more so than in the admittance tests with the same $L_N$ (because no subjective task was required in the admittance tests). Hence, although values of $L_N$ above 110 dB SPL were not generally investigated, we suspect that most subjects would have extreme difficulty in performing the task at these elicitor levels. Some measurements were, however, obtained by S1 (the author) at $f_1$ = 630 Hz for $L_N$ up to 115 dB SPL; these results suggested that transmission-reduction continued to increase at the rate of 0.6 dB/dB beyond the data shown in Fig. 10.

Relevant to (b) are the studies on adaptation and/or relaxation of the acoustic relfex as indicated by the temporal course of eardrum admittance-change. [See Wiley and Karlovich (1975) for a recent summary of this literature.] Although the specific reflex-elicitor employed herein, 2 to 4 kHz noise, has not been used in other studies, we expect that adaptation over the period 5 to 15 sec after elicitor onset is small, especially for $L_N \geqslant 100$ dB SPL. This expectation is based on the adaptation results at comparable elicitor levels (re reflex threshold) reported by Wiley and Karlovich (1975) with wide-band noise and by Dallos (1964) with 2.8-kHz low-pass noise. [In a qualitative supplementary test, we confirmed the lack of adaptation by observing "stable" admittance-changes (visually monitored on a Grason-Stadler 1720 Oto-admittance Meter) over 15 sec for $L_N \geqslant 100$ dB SPL for each subject.

For all the reflex data, the individual-subject data at each test frequency were eyeball-fitted with straight-line segments as in Fig. 10. That is, the data were represented by a growth-rate of attenuation or "regulation efficiency" (Borg, 1972c) of dB of transmission-reduction per dB increase in $L_N$, and a value of $L_N$ at which the straight-line intercepts zero transmission-change. [At 1 kHz, S4 exhibited small transmission-enhancement, and these data were fit with $\Delta|H| = +1$ dB for $L_N \geqslant 100$ dB SPL.] This fitting procedure reflects the fact that the reflex-induced attenuations appear to continue to increase up to the most intense elicitor used. This growth throughout the range of $L_N$ is in contrast to the near-saturation of admittance-change observed for $L_N$ above 105 dB SPL. Of course, as a result of the variability in the transmission-change data and the fact that the transmission-change data were obtained for $L_N$ extending only 5 dB (and sometimes less) above 105 dB SPL, our results do not clearly indicate a continued increase for $L_N \geqslant 105$ dB SPL <u>at the same rate</u> as occurs for $L_N$ below this level.

An alternative hypothesis, namely, that transmission-change is constant for $L_N \geqslant 105$ dB SPL does not fit the data as well as the fit by an increasing attenuation (e.g., see the averages of the data in Fig. 10).

The fitted functions, including a smooth transition section, were averaged across the four subjects at 5 dB increments of $L_N$ from 80 to 110 dB SPL. These averages[5] (see Fig. 11) exhibit the following features:

(1) For $f \geqslant 2$ kHz, no measureable transmission-change occurs for the range of elicitor levels tested.

(2) For $L_N \leqslant 85$ dB SPL, no measureable transmission-change occurs for the range of frequencies tested.

(3) At low frequencies ($f \leqslant 0.63$ kHz)[6], the reflex-induced attenuations are largest and are nearly independent of frequency; the average growth-rate of attenuation is 0.5 dB/dB (except near threshold).

(4) For the maximum elicitor level tested, $L_N = 110$ dB SPL, the average attenuation is about 10 dB for $f \leqslant 630$ Hz, 3.4 dB at 1 kHz, and zero above 2 kHz.

The results for the individual subjects are generally similar (see Fig. 12); for $L_N = 110$ dB SPL, the low-frequency attenuations range from about 8 dB (for S4) to 13 dB (for S2).

## IV. C. Discussion

### (1) Some questions of the interpretation

In applying the $2f_1-f_2$ cancellation-phase method we have assumed that the contralateral noise produces changes in cancellation-phase, $\Theta_c$, only through changes in the magnitude of middle-ear transmission, $\Delta|H(f_1)|$. However, other mechanisms can be suggested by which $\Theta_c$ might be affected.

First, changes in the activity of the olivocochlear bundle (OCB) [see Klinke and Galley (1974) for review] might change $\Theta_c$. It has been demonstrated in laboratory animals that nerve fibers of the crossed and uncrossed OCB respond to moderate- and high-intensity sounds (c.f. Fex, 1962, 1965) and that stimulation of these tracts reduces cochlear outputs (c.f., Galambos, 1956; Desmedt, 1962; Desmedt and LaGrutta, 1963; Wiederhold and Kiang, 1970; Teas et al, 1972). Since the $2f_1-f_2$ generation and cancellation processes most likely occur within the cochlea [see Sec. IV. A. (1)], it is conceivable that a response of the OCB to the eliciting stimulus could affect the $\Theta_c$ measurements. If the site of efferent action is peripheral to the site of the $2f_1-f_2$ generation process, then there is no way to separate middle-ear reflex-effects from efferent effects in these results. However, the effects of the efferent activity may occur central to the $2f_1-f_2$ cancellation process, in which case there may be no effect of efferent activity upon $\Theta_c$. At the present time uncertainty as to the exact sites of efferent action and $2f_1-f_2$ generation and cancellation precludes specification of the order (peripheral vs central) of these mechanisms. This uncertainty coupled with the lack of information of how the OCB responds to acoustic inputs in humans makes it difficult to ascertain the possible involvement of the OCB in our $\Theta_c$ measurements. Nevertheless, in support of the hypothesis that middle-ear transmission-changes solely are responsible for the measured $\Delta\Theta_c$'s, we note that (a) the reflex thresholds for admittance- and transmission-change are nearly equal in our subjects (see Sec. V. A), and (b) the growth properties and frequency dependencies of transmission-change in our subjects are similar to those observed in comparable animal studies (see App. III) where the influence of the OCB might be expected to be small because of the effects of anesthetic agents.

Second, the possible source of changes in $\Theta_c$ is nonlinearities in middle-ear transmission which are introduced by MEM contraction. The post-middle-ear origin of $2f_1-f_2$ for the normal ear has already been discussed [see Sec. IV. A. (1)]; however, what evidence bears on middle-ear linearity during MEM contraction? (1) Wever and Bray (1942), on the cat, have reported intensity functions and some distortion measurements for the cochlear-microphonic-potential vs eardrum sound pressure with (artificially applied) tension on the stapedius tendon as a parameter. Their results are consistent with our assumption that for moderate-intensity stimuli ($P_d \lesssim 80$ dB SPL), MEM contraction does not degrade the incremental linearity of middle-ear transmission. For high stimulation levels, there was some indication that tensioning the stapedius caused increased distortion within the middle-ear. (2) Recent measurements by Nedzelnitsky (1976), on the cat, (which are reviewed in Sec. V.C) are consistent with the incrementally-linear hypothesis, and furthermore, suggest that the linearity of the "reflexed" ear may extend to high stimulation levels. (3) Some indirect support derives from measurements we have made on the effects of earcanal static pressure ($P_{DC}$) on middle-ear transmission. Figure 13 shows a comparison of transmission-change measurements vs $P_{DC}$ obtained using the $2f_1-f_2$ cancellation-phase method with measurements obtained using a pure-tone loudness-balance method. The two procedures yielded nearly equal measures of transmission-change. While the perturbation to the middle-ear system caused by $P_{DC}$ is different from that resulting from the acoustic reflex, we note that the equivalent results between the two psychophysical methods used in the $P_{DC}$ tests were observed over a range of transmission-reductions up to 15 dB in Fig. 13 (and to 20 dB in tests on other subjects) which exceeds those observed in the reflex tests, 8 to 13 dB. To summarize, although none of the arguments given in this paragraph rules out the possibility that increased middle-ear distortion has contaminated our transmission-change estimates, they all indicate that perturbations of the middle-ear can be made without causing enough distortion to influence the measurements reviewed.

### (2) Comparison to other measurements

#### (2.a) Magnitude and frequency dependence of transmission-change

A review of transmission-change measurements from studies on animals is given in App. III. These studies indicate that for a given level of stapedius contraction, the frequency dependence of the induced transmission-changes is similar to that observed in our human data. The transmission-changes are predominantly negative, i.e. attenuation occurs; the largest attenuations occur at low frequencies and are nearly independent of frequency. Contractions causing 5 to 15 dB of attenuation at low frequencies [which are near-maximum (contractions) for guinea-pig, sub-maximum for cat, and probably sub-maximum for rabbit] cause little or no effect (< ±3 dB) at high frequencies. For stronger contractions, but those which are still likely to be within normal physiological range for cat (and perhaps rabbit), attenuation occurs at all frequencies with amplitudes of 20 to 30 dB at low frequencies and 5 to 10 dB at the highest frequencies tested, ~10 kHz. These effects significantly exceed those measured in our tests; however, the maximum capabilities of the human acoustic reflex cannot be deduced from our tests since transmission-change did not saturate for the range of elicitor levels tested.

In human, the difficulty involved in obtaining a direct measure of middle-ear output (without surgical intervention) has limited objective determinations of the changes in transmission occurring during contractions of the middle-ear muscles (MEM). However, several psychophysical methods have been utilized to assess these effects indirectly. With the MEM activated by intense sound, i.e. for acoustic reflexes, measurements have been made of the effects on:

(a) ipsilateral and contralateral remote masking in normal ears (c.f., Burgeat and Hirsh, 1961; Ward, 1961, 1967; Gjaevenes et al, 1969) and comparative measurements on ears with a reduced or non-functioning stapedius reflex (c.f., Fletcher and King, 1963; Harbert and Young, 1965; Bilger, 1966; Smith et al, 1966; Ward, 1967);

(b) temporary threshold shifts in normal ears (c.f., Fletcher and Riopelle, 1960; Ward et al, 1961; Melnick, 1967; Ward, 1967; Karlovich and Wiley, 1974; Morgan and Dirks, 1975) and comparative measurements in ears with a reduced or non-functioning stapedius reflex (c.f., Perlman, 1938; Fletcher and King, 1963; Steffen et al, 1963; Smith et al, 1966; Ferris, 1967; Mills and Lilly, 1970);

(c) loudness contours for high-intensity tones (Reger, 1960; Ross, 1968a, 1968b; Morgan et al, 1974); and

(d) pure-tone absolute thresholds and suprathreshold loudness judgements (Shapley, 1954; Kirikae, 1960; Loeb and Riopelle, 1960; Reger, 1960; Loeb and Fletcher, 1961; Prather, 1961; Ward, 1961, 1967; Vigran, 1965; Morgan and Dirks, 1975).[7]

The effects of acoustic-stapedius reflexes on middle-ear transmission have also been assessed indirectly, but objectively, on patients having peripheral unilateral facial paralysis (Borg, 1968; Borg and Zakrisson, 1974).

Within this literature inconsistencies exist as to the qualitative effects of acoustic reflexes on middle-ear transmission, and methodological problems with some of the studies have been identified (c.f., Loeb, 1962; Mills, 1969; Ward, 1974; Morgan and Dirks, 1975). For the present discussion [and Sec. IV. B. (2.b)] we shall focus on comparison of our results with the objective measurements reported by Borg (1968; Borg and Zakrisson, 1974); however, we note that for most of the other studies for which non-zero effects on transmission were measured, the frequency dependence of the transmission-changes is generally similar to that indicated by our results. [In Sec. V. C we consider the hypothesis deriving from some of the studies in (d) that a given MEM contraction reduces the transmission of high-intensity stimuli more than near-threshold and moderate-intensity stimuli.]

In 1968, Borg reported on an objective procedure providing for a determination of transmission-change induced by the acoustic stapedius reflex. The procedure made use of subjects with a temporary loss of stapedius-muscle function in one ear arising from peripheral unilateral facial paralysis of short duration (Bell's palsy). During the paralysis and following recovery, (relative) acoustic impedance-change (at 800 Hz) was measured on the normal ear as a function of the sound-pressure level of tones presented to the (contralateral) affected ear. The differences in the sound-pressures required to elicit the same impedance-changes were taken as estimates of the transmission-change provided by the stapedius reflex functioning in the recovered ear. In this method two principal assumptions are made: First, it is assumed that impedance-change in the normal ear is a fixed single-valued function of the contralateral middle-ear output, whether or not the stapedius muscle is paralyzed. [This function may depend on the frequency of the eliciting stimulus.] Second, it is assumed that Bell's palsy eliminates the ability of the stapedius muscle to cause transmission-change on the affected ear, but in no other way influences the relation between middle-ear output and the strength of muscle contraction.

Borg (1968) obtained data on four subjects using elicitor frequencies (in the affected ear) of 0.5, 1.45, 2, and 3 kHz. At 500 Hz, tones with an intensity of 20 dB above the reflex-threshold were attenuated 12 to 15 dB. The growth of attenuation was approximately linear with elicitor intensity; hence, the results are summarized by a "regulation efficiency" (Borg, 1972c) of 0.67 dB/dB [averaging 12/20 and 15/20]. At 1.45 kHz, attenuations of 0 to 6 dB were observed (for tones +16 dB re reflex-threshold), and lesser effects occurred at 2 and 3 kHz. As one component of a later study Borg and Zakrisson (1974) made similar measurements on 19 subjects at 500 Hz. A greater range of elicitor intensity was examined, up to 30 dB above the reflex-threshold for which attenuations occurred; the mean data show a maximum attenuation about 20 dB and are well fit by the growth-rate of attenuation of 0.67 dB/dB.[8]

The frequency dependence of our transmission-change results is in close agreement with Borg's results. The maximum attenuation of ~20 dB (at 500 Hz) indicated by Borg and Zakrisson significantly exceeds our maximum low-frequency attenuation of ~10 dB; their results suggest that the human acoustic-stapedius reflex is capable of reducing middle-ear transmission by nearly the amount noted earlier to have been measured in cat, 20 to 30 dB.

### (2.b) Dependence of transmission-change on elicitor level

In Borg's (1968; Borg and Zakrisson, 1974) tests and in ours reflex-induced attenuation (in dB) measured at low frequencies grows roughly in proportion to elicitor level (in dB re reflex threshold). The growth-rates of attenuation are 0.67 dB/dB in Borg's data and 0.5 dB/dB in ours. Because experimental conditions for the studies are different, analysis of the differences is necessary before making comparisons of the growth-rates. To aid in this analysis some variables of interest are specified in Fig. 14.

In Borg's paradigm, transmission-changes were imposed on 500-Hz tones directly as a consequence of the ipsilateral reflexes that the tones themselves elicited. Hence, Borg's growth-rate of attenuation, 0.67 dB/dB, indicates the dB reduction in transmission (H) per dB increase in input sound-pressure ($P_d$) delivered to the regulating ear, i.e. the measured quantity was $\eta_i = -\Delta\log|H_R(500\text{ Hz})|/\Delta\log|P_{dR}(500\text{ Hz})|$ where $\eta$ denotes growth-rate of attenuation and the subscript "i" indicates that the growth-rate is referred to changes in middle-ear input (and where we have assumed the Right ear was the affected ear). Another quantity can be defined which also summarizes Borg's results: $\eta_o = -\Delta\log|H_R(500\text{ Hz})|/\Delta\log|U_{sR}(500\text{ Hz})|$ where the subscript "o" indicates that the growth-rate is referred to changes in middle-ear output. Because $H = U_s/P_d$, it follows that $\eta_o = \eta_i/(1-\eta_i)$. Hence, $\eta_i = 0.67$ dB/dB implies that $\eta_o = 2$ dB/dB; in words, each dB increase in middle-ear output is accompanied by a 2 dB reduction in middle-ear transmission. [The quantities $\eta_i$ and $\eta_o$ are denoted by Borg (1972c), respectively, as the "closed-loop" and "open-loop" "efficiencies of regulation".]

In our paradigm, transmission-change was measured at ~500 Hz (400, 500, and 630 Hz) as in Borg's tests, however, the reflex-eliciting stimulus was delivered to the contralateral ear and was composed of high frequencies, 2 to 4 kHz noise. Hence, our growth-rate of attenuation of 0.5 dB/dB = $-\Delta\log|H_R(\sim 500\text{ Hz})|/\Delta\log|P_{dL}(2\text{ to }4\text{ kHz})|$.

Thus, in order to (meaningfully) compare our measured $\Delta\log|H_R(\sim 500\text{ Hz})|$ for our elicitor presented at a given level (re reflex threshold) to Borg's measured $\Delta\log|H_R(500\text{ Hz})|$ for Borg's elicitor presented at the same level (re its reflex threshold) we must first consider whether the two elicitors can be expected to activate the same MEM response, $M_R$, since it is a given $M_R$ which we assume to cause a given $\Delta\log|H_R(500\text{ Hz})|$. That is, we need to determine the relationship between the growth of $M_R$ vs $P_{dL}(2\text{ to }4\text{ kHz})$ and the growth of $M_R$ vs $P_{dR}(500\ \mathrm{Hz})$.

Before considering this relationship in human, we examine measurements in animals of the growth of transmission-change vs elicitor level for different eliciting stimuli. Borg (1972a, 1972c) measured acoustic-reflex regulation in rabbit using two procedures (see App. III). In the 1972a paper, an experiment functionally equivalent to ours was performed: measurements were made of the change in cochlear-microphonic-potential of a moderate-intensity 800-Hz tone vs the intensity of a 2-kHz contralateral elicitor. This experiment indicated a growth-rate of attenuation of 0.43 dB/dB which is similar to the 0.5 dB/dB value indicated by our tests. In the 1972c paper, the indirect method used in human [Borg, 1968; see Sec. IV. C. (1)] was applied to the rabbit, except that the action of the stapedius muscle in the eliciting ear was removed by transection in the rabbit rather than by palsy as in the human. These tests indicated that $\eta_i = 0.7$ dB/dB ($\eta_o = 2.3$ dB/dB) which is similar to Borg's (1968; Borg and Zakrisson, 1974) value in human. Thus, the difference in growth-rate of attenuation shown in human between our data and Borg's ~~are~~ is also seen in results from experiments in rabbit with similar differences in paradigm.

As noted above, two differences existed between the elicitors used in our tests and Borg's tests in human (and between Borg's two tests in rabbit): (1) ours was contralateral and his was ipsilateral (to the measured-$\Delta|H|$ ear), and (2) the elicitor frequencies differ~~r~~ed.

The influence of these two differences on the growth of MEM response vs elicitor level [i.e. the differences between $M_R$ vs $P_{dL}$(2 to 4 kHz) and $M_R$ vs $P_{dR}$(500 Hz)] are considered separately below. To determine these relationships, we make use of the assumption that a given $M_R$ implies a given admittance-change, $\Delta Y_{dR}$ (at a particular frequency), and we consider reported measurements of the growth of $\Delta Y_{dR}$ (as a relative index of $M_R$) vs the levels of different reflex elicitors which are comparable to ours and Borg's (in human). We consider the growths of the functions M vs $P_d(f)$ in terms of (a) the slopes $\Delta M/\Delta P_d(f)$ of their (approximately) linear regions [where $\Delta P_d(f)$ is assumed to be in dB], and/or (b) the $\Delta P_d(f)$'s required to produce the same $\Delta M$.

(1) In human, some measurements comparing the growth of ipsilateral MEM response [$M_R$ (as indicated by $\Delta Y_{dR}$) vs $P_{dR}(f)$] to the growth of contralateral MEM response [$M_R$ vs $P_{dL}(f)$] have been reported by Møller (1961b, Figs. 2, 3, and 6; 1962a, Fig. 2; 1962b, Figs. 3, 6, 8, and 9). There is some variability in the slopes of the functions $M_R$ vs $P_{dR}(f)$ and $M_R$ vs $P_{dL}(f)$, and no high-frequency elicitors were tested ($f \lesssim 1500$ Hz); nevertheless, the ipsilateral growth is generally observed to be "somewhat steeper" (Møller, 1961b, pg.743) than the contralateral growth. In rabbit, measurements of $M_R$ (as indicated by $\Delta Y_{dR}$) vs $P_{dR}(f)$ and $P_{dL}(f)$ for low and high values of f have been reported by Borg (1972b, Figs. 2 and 6); the ipsilateral growth is slightly steeper than the contralateral growth. Therefore, we expect that Borg's measured values of $\eta_i = 0.67$ dB/dB in human (Borg, 1968;

Borg and Zakrisson, 1974) and 0.7 dB/dB in rabbit (Borg, 1972c) would decrease somewhat if the 500-Hz elicitor were applied to one ear and transmission-change at 500 Hz were measured in the contralateral ear [using a method (such as ours) involving sub-reflex-eliciting test stimuli]. In the following discussion we will assume that the ipsilateral vs contralateral difference in growth-rate is rather small and independent of elicitor frequency, and we will ignore it while considering the effects of elicitor frequency.

(2) An influence of elicitor frequency, f, on M vs $P_d(f)$ might be expected to occur because the frequency dependence of $\Delta H(f)$ indicates that low-frequency ($f_{LO}$) elicitors are attenuated by MEM contraction more than high-frequency ($f_{HI}$) elicitors. However, the response of the nervous system [i.e. M vs $U_s(f)$] may also depend on elicitor frequency.

In rabbit, measurements of MEM response (as indicated by $\Delta Y_d$) for low- and high-frequency (ipsilateral and contralateral) elicitors, i.e. M vs $P_d(f_{LO})$ and M vs $P_d(f_{HI})$, have been reported by Borg (1972b, Figs. 2 and 6). The growth of M vs $P_d(f_{LO})$ is significantly <u>steeper</u> than the growth of M vs $P_d(f_{HI})$; e.g. $\Delta P_d(f)$ required to change M from 10 to 80% of its maximum value is about 15 to 20 dB at 500 Hz, but is about 30 dB for f = 2 or 4 kHz. Hence, for $\Delta P_d(f_{LO}) = \Delta P_d(f_{HI})$, (a) $\Delta M$ for $f_{LO} > \Delta M$ for $f_{HI}$, and (b) $\Delta U_s(f_{LO}) < \Delta U_s(f_{HI})$ [because of the frequency dependence of $\Delta H(f)$]; it follows that the response of the nervous system, M vs $U_s(f)$, is significantly steeper for low frequencies than for high frequencies. [Borg (1973, pg. 17) reaches a similar conclusion.] Furthermore, the two measures of the growth-rate of attenuation referred to input-sound-pressure change, 0.43 dB/dB for a high-frequency elicitor (Borg, 1972a) and 0.7 dB/dB for a low-frequency elicitor (Borg, 1972c), are in nearly the same ratio ($0.43/0.7 \simeq 0.6$) as are the relative growths of M vs $P_d(f_{HI})$ and M vs $P_d(f_{LO})$ (Borg, 1972b), so that these two regulation measures are in agreement with respect to the differences associated with elicitor frequency.

In human, measurements of MEM response (as indicated by $\Delta Y_d$) for low- and high-frequency elicitors have been reported in several studies (see below). Most of these measurements are for contralateral reflex activation, i.e. results are reported for $M_R$ vs $P_{dL}(f)$. The effect of elicitor frequency in these measurements can be considered conveniently in terms of the quantity $\Psi = \Delta P_{dL}(f_{LO};\Delta M_R)/\Delta P_{dL}(f_{HI};\Delta M_R)$ where $\Delta P_{dL}(f;\Delta M_R)$ is the dB change in sound pressure required to elicit a given change in contralateral MEM response, $\Delta M_R$.

Functions for $M_R$ vs $P_{dL}(f)$ for various low- and intermediate-frequency elicitors (i.e. $f \leqslant 600$ Hz and $1000 \leqslant f \leqslant 1500$ Hz) are given by Møller (1962a, Fig. 3; 1962b, Figs. 3 and 9) and Hung and Dallos (1972, Fig. 3). These functions indicate generally similar growths, with some tendency for steeper growth for intermediate-frequency elicitors; i.e. they indicate values for $\Psi$ in the range of about 0.8

to 1.5. Møller (1962a, Figs. 1 and 4) also shows contours of constant contralateral impedance-change for single-tone elicitors from 200 to 4000 Hz. These contours indicate that the $\Delta P_{dL}(f)$ required to change $M_R$ from 10 to 70 or 80% of its maximum value is roughly similar for elicitors in the range of 300 to 500 Hz and in the range of 2 to 4 kHz, i.e. $0.8 \leqslant \Psi \leqslant 1.5$.

Further evidence on $M_R$ vs $P_{dL}(f)$ derives from three studies in which "reflex-growth" was measured using elicitors ranging from 250 Hz to 4 kHz and on relatively large subject sets, N = 10 to 45 (see Fig. 15). The results from Djupesland et al (1967) and Peterson and Lidén (1972) indicate that (a) the growth of $M_R$ vs $P_{dL}(f)$ is less steep for a 4-kHz elicitor in comparison to that which occurs for $f \leqslant 2$ kHz, and (b) the growths for $f \leqslant 2$ kHz are generally similar (i.e. within a ratio of 0.8 to 1.3). The results from Beedle and Harford (1973) exhibit growths of $M_R$ vs $P_{dL}(f)$ which increase with frequency, with the growth for the 2-kHz elicitor being about 1.5 times the growth for the 0.5-kHz elicitor. Thus, the results from Beedle and Harford suggest $\Psi \simeq 1.5$ whereas the other two studies in Fig. 15 suggest $0.8 \lesssim \Psi \lesssim 1.3$.

The above results in human indicate values for $\Psi$ generally near 1. Insofar as $\Psi = 1$, there is <u>no</u> effect of elicitor frequency on $M_R$ vs $P_{dL}(f)$; this implies that (a) our measured growth-rate of attenuation of 0.5 dB/dB should be compared directly to Borg's (1968; Borg and Zakrisson, 1974) $\eta_i = 0.67$ dB/dB, in which case these values are in reasonably good agreement [particularly if we allow for the somewhat steeper growth of ipsilateral vs contralateral MEM response (as noted above)], and (b) in human (as in rabbit), the response of the nervous system, M vs $U_s(f)$, is steeper for low frequencies than for high frequencies.

In summary, a framework has been given for considering quantitative relationships between elicitor frequency and level, and the growths of MEM response, transmission-change, and (relative) admittance-change. Within this framework, the two values of the growth of attenuation vs elicitor level obtained in human by us and Borg (1968; Borg and Zakrisson, 1974) can be compared, and so can the two values obtained in rabbit by Borg (1972a, 1972c). In rabbit, available data suggest that the nervous-system response vs middle-ear output [M vs $U_s(f)$] exhibits a strong dependence on frequency [with $M(U_s(f_{LO})) >> M(U_s(f_{HI}))$] and that the two regulation measures obtained by Borg (1972a, 1972c) are reconcilable when elicitor-frequency differences are accounted for. In human, a similar frequency dependence for M vs $U_s(f)$ may exist, in which case our regulation value and Borg's (in human) are in agreement when elicitor-frequency (and ipsilateral vs contralateral) differences are accounted for.

## V. Relations between admittance- and transmission-change

### V. A. Basic results

Our average results for $\Delta|Y_d(f)|$ and $\Delta|H(f)|$ for $L_N$ = 110 dB SPL are plotted together for comparison at the top of Fig. 16. For the cat and rabbit, Møller (1965) has also compared $\Delta|Y_d(f)|$ and $\Delta|H(f)|$ resulting during stapedius contraction and his results, each for one animal, are also given in Fig. 16. [Further details about Møller's experiments are given in App. III.] Note first that the form of the frequency dependence for the three $\Delta|H(f)|$ functions are quite similar, and so to are those for $\Delta|Y_d(f)|$ except for about a one-half-octave upward shift for the $\Delta|Y_d(f)|$ data for the cat and rabbit relative to our results for human. Additionally, note that the functions for $\Delta|H(f)|$ are generally below those for $\Delta|Y_d(f)|$. The difference at low frequencies is large in our results for human, $\Delta|H| \approx -10$ dB while $\Delta|Y_d| \approx -2.5$ dB, and appears small in cat.

The growth properties of admittance- and transmission-change with elicitor level can also be compared. For our results, admittance-change exhibited a limited dynamic range, nearly saturating for $L_N \geqslant 105$ dB SPL. In contrast, transmission-reduction continued to increase up to $L_N$ = 110 dB SPL. For comparison, Borg (1972a), on the rabbit anesthetized "lightly" with sodium pentobarbital, measured (relative) impedance-change and transmission-change at 800 Hz, both as functions of the level of a 2-kHz contralateral elicitor. Impedance-change nearly saturated for elicitor levels above 110 to 115 dB SPL. In contrast, transmission-reduction continued to increase up to the maximum elicitor level tested of 125 dB SPL. These results are in agreement with ours.

With respect to the reflex-thresholds, Borg (1972a) noted, from his measurements in rabbit and decerebrate cat, that the thresholds for impedance- and transmission-change were about equal, within 2 to 3 dB. For our results, the reflex-thresholds were defined rather coarsely; but consistent with Borg's findings, the average thresholds for admittance- and transmission-change were nearly equal at $L_N \simeq 85$ to 90 dB SPL. Furthermore, this correspondence is also evident in our individual subject data. From Fig. 9 observe that the threshold for transmission-change for S2 is $L_N \simeq 87$ dB SPL, whereas the thresholds for S1, S3, and S4 are higher at $L_N \simeq 91$ to 94 dB SPL. Similarly, from Fig. 1 the threshold for admittance-change for S2 is again lowest at $L_N \simeq 85$ dB SPL with the remaining subjects' (estimated) thresholds at $L_N \simeq 90$ dB SPL.[9]

We now attempt to interpret the acoustic-reflex effects on admittance and transmission jointly in terms of changes in specific elements of the middle-ear system.

## V. B. Implications for the mechanics of the middle-ear

Because at low frequencies the reduction in transmission is larger than the reduction in admittance for a given stapedius contraction, Møller (1965) concluded that the middle-ear system (including the cochlear load on the stapes) could not be represented as a "series" circuit in both the resting and reflexed states. In other words, because $\Delta|H(f)| < \Delta|Y_d(f)|$, the volume velocities of the eardrum $[U_d(f)]$ and the stapes $[U_s(f)]$ cannot be equal (or proportional) to each other in both the resting and reflexed states; hence, a portion of $U_d(f)$ does not appear as $U_s(f)$, i.e. some "shunting" of volume velocity occurs.[10] The fact that with increasing elicitor level, admittance-change exhibits near-saturation prior to transmission-change is also consistent with the view that shunting occurs within the middle-ear system. The result in Fig. 16 indicating that (at low frequencies) the difference between $\Delta|Y_d(f)|$ and $\Delta|H(f)|$ is larger in human than in cat, for a condition in which the $\Delta|H(f)|$'s are about the same, is consistent with the hypothesis that in the resting middle-ear more shunting occurs in human than in cat.[11]

The concepts in the above paragraph can be illustated more specifically with the aid of the circuit models proposed by Wzislocki (1962) for the human middle-ear, see Fig. 17, and by Peake and Guinan (1967) for the cat middle-ear, see Fig. 18. The effects of stapedius contraction were not explicitly included in these models. However, as a first-order approximation we suggest that the principal effect of stapedius contraction is to cause a decrease in the compliance of the elements $C_c$ and $C_2$ respectively in the two models. This decrease is meant to reflect (a) increased (incremental) stiffness provided directly from the tensioned stapedius muscle, and (b) increased stiffness of the annular ligament (supporting the footplate of the stapes in the oval window) which may be stretched as the muscle contraction displaces the stapes from its normal position. In Fig. 19 we show predictions for $\Delta|H(f)|$ and $\Delta Y_d(f)$ caused by (arbitrarily chosen) successive reductions of these compliance elements. [Ross (1968a) has performed some similar calculations with the Zwislocki model.] Our data in human and Møller's in cat are included for comparison. Note that the main features of the frequency dependencies in the data including the sizes of $\Delta|H(f)|$ and $\Delta|Y_d(f)|$ at low frequencies and the maximum $\Delta \measuredangle Y_d(f)$ at intermediate frequencies are reasonably fit by the four-fold reductions in $C_c$ or $C_2$.

For the Peake-Guinan cat-model with $C_2$ reduced by four, the transmission-reduction at low frequencies is only slightly larger than the admittance-reduction, i.e. $\Delta|H| = -7.8$ dB and $\Delta|Y_d| = -6.6$ dB. This occurs because shunting in the Peake-Guinan cat-model is so small, $C_j << C_2$. Thus, reducing $C_2$ from its normal value at first causes nearly equivalent changes in admittance and transmission. With larger reductions, however, the shunt current through $C_j$ becomes significant in limiting admittance-change, and the low-frequency transmission-reduction then exceeds the admittance-reudction. As $C_2 \rightarrow 0$ transmission decreases proportionately with $C_2$-reduction, but admittance-reduction saturates (not shown in Fig. 16) at $\Delta|Y_d| \simeq -20$ dB.

For the Zwislocki human-model, all reductions of $C_c$ cause much larger transmission-reductions than admittance-reductions, e.g. for $C_c$ reduced by four $\Delta|H| \simeq -10$ dB but $\Delta|Y_d| \simeq -3$ dB. This occurs because in the model the shunt currents through the uncoupled portion of the eardrum and the incudo-stapedial joint sum to be a substantial portion of the input current $U_d$. As $C_c \rightarrow 0$ transmission decreases proportionately with $C_c$-reduction, but admittance-reduction saturates at $\Delta|Y_d| \simeq -5$ dB.

While reduction of only the single compliance elements $C_c$ and $C_2$ in the two models reasonably accounts for the main features in the above data, more complex effects of stapedius contraction are suggested by more detailed examination of the above data and by consideration of our individual data.

For human: (1) Observe that the saturation values of admittance-change predicted for the Zwislocki human-model are somewhat larger than those indicated by our average data. For some of our individual data (see Figs. 1, 2, and 12), this discrepancy is larger; e.g. subjects S3 and S4 show low-frequency transmission-reductions of 8 to 10 dB but admittance-reductions (which are "completely" saturated for S4) of only 1 to 1.5 dB. (2) Observe that the near-saturation of admittance-change in the model occurs only when transmission-reduction exceeds 20 dB, whereas our results indicate near-saturation of admittance-change occurs for transmission-reductions of only 10 dB. Of course, the ability of the stapedius contraction to "reduce $C_c$" may be limited, and consequently, transmission-reduction may saturate between, perhaps, 10 and 20 dB. [Attenuation of 20 dB is shown by Borg and Zakrisson, 1974.] In this case, admittance-change would saturate at smaller values more comparable to our average data; however, admittance-change saturation would then be directly related to transmission-change saturation in contrast to the observed data. The above noted discrepancies [(1) and (2)] can be diminished if in addition to causing $C_c$-reduction, stapedius contraction simultaneously causes as increase in shunting (i.e. an increase in shunt admittance) elsewhere in the Zwislocki human-model. Alternatively, the effects of stapedius contraction may be more complicated.

For the cat we are unaware of any measurements on the saturation of admittance-change vs transmission-change. [Møller's (1965) measurements were made for only one level of stapedius contraction, and a level which is clearly sub-maximum with respect to transmission-change.(see App. III).] In the predictions for the Peake-Guinan cat-model, admittance-reduction saturates at about 20 dB, and with transmission-reductions which are appreciably larger. In real ears smaller values may occur; a shunt mechanism may exist such that as "$C_2$" decreases, the shunt-path admittance increases. In the model, the only mechanism which exists to enable "added shunting" would be for $C_j$ to increase as $C_2$ decreases; alternatively, strong stapedius contraction might introduce a shunt path not represented in the Peake-Guinan model. In addition to the above hypothetical arguments, other reported measurements of $\Delta|H(f)|$ in cat provide evidence that $C_2$-reduction alone is inadequate to "explain" the effects of strong stapedius contraction. Specifically, measurements of $\Delta|H(f)|$ for different levels of stapedius contraction (see App. III; Wever and Bray, 1942; Teig, 1973) show that $\Delta|H(f)|$ functions are relatively smooth vs frequency, with those having low-frequency reductions of 20 to 30 dB decreasing to 10 to 15 dB reductions near 2 kHz and 5 to 10 dB reductions near 10 kHz. In contrast, the predictions in Fig. 19(b) based on $C_2$-reduction alone, indicate that increasing low-frequency transmission-reductions are accompanied by increasing intermediate- and high-frequency transmission-changes varying rapidly with frequency. The added shunt admittance that we have hypothesized to occur during strong stapedius contraction may also result in smoothing the $\Delta|H(f)|$ functions thereby diminishing this discrepancy.

For the rabbit the information available on the middle-ear is incomplete, and to our knowledge, no unified description or circuit model has been published. However, some of the measurements in Møller (1965) and Borg (1972a) suggest that the resting middle-ear system may have little shunting. To this extent, (a) the results on the effects of stapedius contraction indicating that admittance-change saturates once transmission-reduction reaches 10 to 15 dB and that transmission-reduction continues to be increasing (Borg, 1972a), and (b) the results in Fig. 13 (from Møller, 1965) indicating that transmission-reduction of 13.5 dB is coincident with admittance-reduction of only 7.5 dB are suggestive that for this species also some increased shunting occurs in the middle-ear during strong stapedius contraction.

In summary, in attempting to relate our results in human on $\Delta Y_d(f)$ and $\Delta|H(f)|$ which occur during acoustic reflexes and corresponding results in cat and rabbit, we have discussed three factors: (1) the amount of shunting present in the resting middle-ear systems of the thre species, (2) the effects of increased stiffness caused by stapedius contraction, and (3) an increased shunting mechanism caused by strong stapedius contraction. For low- and moderate-level acoustic-reflexes, i.e. those causing transmission-reductions less than ~10 dB and non-saturating admittance-changes, factors (1) and (2) appear adequate to account for the observed effects on admittance and transmission.

## V. C. For a given level of MEM contraction, dependence of acoustic-reflex effects on measurement stimulus intensity

In this section we consider a hypothesis concerning acoustic-reflex effects which derives from some studies (in human) on absolute threshold changes and suprathreshold loudness changes. In 1960, Loeb and Riopelle reported a series of psychophysical experiments on threshold shifts and suprathreshold loudness judgments during presumed MEM contraction, as elicited by intense contralateral acoustic stimulation. From these tests the authors suggested (Loeb and Riopelle, 1960, pg. 606): "the relfex, once activated, acts in such a way as to attenuate loud sounds considerably while providing little or no attenuation for sounds near threshold". Ward (1967, 1974) reached this same conclusion. Recently, Morgan and Dirks (1975) reviewed these studies and identified some methodological problems which cast doubt on the above hypothesis. However, on the basis of additional experimentation, Morgan and Dirks (1975) arrived at a similar conclusion; namely, they reported that for a given level of a contralateral reflex-eliciting stimulus, "the effect on loudness increased as the intensity of the loudness-judgment stimulus increased".

Morgan and Dirks (1975) examined loudness-judgment comparisons between a <u>reference</u> tone-pulse presented during a presumed condition of no MEM contraction and a <u>test</u> tone-pulse presented during a presumed condition of different levels of MEM contraction, as elicited by intense contralateral noise, 2.9 to 5.8 kHz, at levels up to 115 dB SPL. Both tone-pulses were brief, 50 msec in duration, which was presumed short enough to prevent any effect on tone transmission by MEM contractions induced by the tones themselves. The results showed that low-frequency, high-intensity reference tones, 250 and 500 Hz at levels of 100 and 115 dB SPL, were matched to test tones of greater intensity (with the matching differences at reference intensity of 115 dB SPL somewhat exceeding those at 100 dB SPL). For example, a 500-Hz reference stimulus at 100 dB SPL was matched in loudness by a test stimulus at 102.7[12] dB SPL in the presence of a 105 dB SPL contralateral elicitor. However, for moderate-intensity reference stimuli, 60 dB SPL, no matching differences occurred. That is, no effect of MEM contraction on loudness judgments of moderate-intensity stimuli was found.

In contrast, our measurements have been made with moderate-intensity stimuli [$P_d$ < 80 dB SPL]; and, (a) transmission-reductions of ~10 dB were observed at low frequencies for the maximum elicitor level tested, and (b) the thresholds, growths, and sizes of admittance- and transmission-change have been related (see Sec. V. B) suggesting that both phenomena ($\Delta Y$ and $\Delta H$) occur together. Also, measurements on animals by Wever and Vernon (1955) and Price (1966) [see App. III] indicate that significant transmission-changes occur for moderate-intensity stimuli during MEM contractions elicited by inetnse contralateral sounds.

Morgan and Dirks state their conclusions for the effects of MEM contraction, cautiously, in terms of the measured psychophysical variable - loudness judgment. However, if their loudness-judgment results are interpreted as being equal to middle-ear transmission-changes (as Loeb and Riopelle suggested), then a direct conflict exists with our results. Their results would then imply what might be called a "strong form" of the Loeb and Riopelle hypothesis; namely, that no significant transmission-change occurs for low- and moderate-intensity stimuli when acoustic-reflex-induced admittance-change is near-maximum. We suggest this hypothesis is invalid, and the following analysis outlines why the results from Morgan and Dirks need not, in fact, be interpreted as support for such a hypothesis.

First, some questions arise concerning the experimental parameters used by Morgan and Dirks. The most intense elicitor (115 dB SPL) was used only with reference stimulus tones at 115 dB SPL; this condition resulted in the largest loudness changes observed, ~6 dB at 500 Hz. Measurements with reference tones at lower levels would have been particularly relevant to the conclusion that the effect of a given MEM contraction inetracts with the level of the loudness-judgment stimuli. Additionally, such measurements in conjunction with variations in the temporal parameters of the experiments (which were done at lower elicitor levels) would have provided stronger evidence to rule out possible undesirable interactions (e.g., binaural loudness summation, contralateral remote masking, and/or backward masking) between the eliciting stimulus and the loudness-judgment stimuli.

Second, we suggest that the psychophysical variable of loudness-judgment may not be uniquely related to middle-ear output. In considering the conclusion reached by Loeb and Riopelle, Wever (1962, pg. 229) states: "This idea is contrary to many previous observations", i.e. measurements of transmission-change in animals, and "Probably, the central appreciation of loudness takes account of the presence of the protective reflexes." A similar explanation was offerred by Carmel and Starr (1963, pg. 613): "It may be that the central processes shown to control middle-ear muscle actions are integrated into sensory systems in such a way as to enable a perceptual differentiation between internal and external modifications of cochlear activity and to compensate at perceptual levels for the internally controlled changes." We agree with these authors and suggest that for some psychophysical judgments the central nervous system accounts for the signals being sent to, and the effects produced by, the contracting MEM. In considering our measurements in this regard, our procedure is based on intracochlear cancellation of the $2f_1-f_2$ combination-tone and is presumably a measure of middle-ear output changes.

The arguments above should not be taken to imply that transmission-change is completely independent of measurement-stimulus level. For a given level of MEM contraction, transmission-reductions may increase with measurement-stimulus level because of middle-ear nonlinearities (in both the resting and reflexed states). At present, data do not exist in human to evaluate this dependence of transmission-change on level. Some recent measurements on anesthetized cats by Nedzelnitsky

(1976) are, however, relevant. As a function of eardrum sound-pressure-level ($P_d$) and tone frequency, measurements were made of the fundamental component (i.e. through a 3-Hz tracking filter) of the sound pressure developed in scala vestibuli of the basal turn of the cochlea ($P_c$).[13] The intracochlear sound pressure can be regarded as a relative measure of middle-ear output, just as is (round-window) cochlear-microphonic-potential, but with the advantage that it retains linearity at high stimulation levels. Measurements were made under two conditions, with and without tetanic electric stimulation applied to the stapedius muscle (with the bulla open), and transmission-change was computed as $\Delta|H(f)| = \Delta|P_c(f)/P_d(f)|$. The resulting stapedius contractions were strong; transmission-reductions of 20 to 30 dB occurred at low frequencies, which are similar to results reported in other comparable studies on cat [see App. III; Wever and Bray, 1942; Teig, 1973]. Furthermore, the attenuations measured for $P_d$ = 85, 110, and 135 dB SPL were nearly equal (in dB); i.e. no excessive attenuations occurred for the 135 dB input-tones relative to the attenuations observed for the lower-level stimuli. Thus, at least in the cat, significant stapedius contraction left the middle-ear system as an approximately incrementally-linear system throughout the range of stimulation level studied.

## VI. Summary

As a function of the level $L_N$ of a contralateral acoustic-relfex elicitor, admittance- and transmission-change exhibited approximately equal "reflex thresholds" at $L_N$ = 85 to 90 dB SPL; however, for $L_N \geq 105$ dB SPL, admittance-change nearly saturated whereas transmission-change continued to increase. For $L_N$ = 110 dB SPL, the average low-frequency admittance-change was -2.5 dB; the average low-frequency transmission-change was -10 dB.

The above results and corresponding results in animals were interpreted in terms of changes in the linear-incremental properties of the mechanics of the middle-ear systems of the different species. Our analysis suggested that increased stiffness caused by stapedius contraction reasonably accounts for the effects on admittance- and transmission-change of low- and moderate-level acoustic reflexes.

In attempting to compare the growth of low-frequency transmission-change vs elicitor level indicated by our and Borg's (1968; Borg and Zakrisson, 1974) tests in human, and also comparing the values indicated by Borg's (1972a, 1972c) two tests in rabbit, we emphasized the different forms of regulation measures and the importance of the different elicitor-frequency compositions. A framework was presented for considering quantitative relationships between elicitor frequency and level, and the growths of MEM response, transmission-change, and (relative) admittance-change. Available data suggested that for human and rabbit, (a) the response of the nervous system to changes in middle-ear output is a steeper function for low frequencies than for high frequencies, and (b) the reported regulation measures are in reasonable agreement when elicitor differences (frequency, and ipsilateral vs contralateral) are accounted for.

Finally, our data (and other data in animals) show that acoustic reflexes change middle-ear transmission for moderate-level stimuli. Insofar as other psychophysical experiments have indicated little effect of MEM contraction on transmission of low- and moderate-level stimuli, we suggest that for some psychophysical judgments the nervous system may account for the signals being sent to, and the effects produced by, the contracting MEM.

## Footnotes

1. Note that no accepted standard exists for coupler calibration of circumaural earphone/housing systems. Some of the problems associated with such calibrations (e.g. uncontrolled resonances above ~2 kHz and uncontrolled leakage influencing calibration below ~500 Hz) are well known, and some solutions to the calibration problem have recently been suggested (c.f., Charan et al, 1965; Benson et al, 1967; Villchur, 1970; Zwislocki, 1971; Shaw, 1974).

2. It may be of interest that the absolute sensitivity measured for both earphone/housing systems on the Extended NBS-9A coupler was 106 dB SPL/dBV.

3. Note that the prefix $\Delta$ indicates a difference, i.e. $\Delta R_d$ is $R_d$ with the reflex active minus the resting $R_d$, whereas earlier (Figs. 1 and 2) the prefix $\Delta$ was used to indicate a ratio.

4. For noise elicitors with a 2-kHz bandwidth (logarithmically) centered at 2 and 4 kHz, Flottorp et al (1971, Fig. 2) report mean thresholds of 80 and 90 dB SPL (average = 85 dB SPL), and Popelka et al (1976, Fig. 2) report mean thresholds of 83 and 89 dB SPL (average = 86 dB SPL). For a noise elicitor high-passed above 2.6 kHz [and which for the TDH-39 earphone and MX-41/AR cushion used is reported (Popelka et al, 1974) to have an upper

frequency limit of ~5.8 kHz], Margolis and Fox (1976, Table 2) report a mean threshold of 81.3 dB SPL. Although this elicitor is of larger bandwidth than ours which would tend to give it a lower threshold, it is also of higher center frequency which tends to offset the bandwidth advantage. Finally, for noise elicitors with a 600-Hz bandwidth centered at 2 and 4 kHz, Peterson and Lidén (1972, Table 2) report mean thresholds of 76.2 and 82.4 dB SPL (average = 79.3 dB SPL). Because these elicitors are of smaller bandwidth than ours, their thresholds are expected to be above ours by 0 to 2 dB.

5. Additional data are given in Appendix I.

6. The lack of data below 400 Hz stems from a failure of the psychophysical method. At lower frequencies the CTC becomes perceptually "muddy" and the cancellation-phase nulling accuracy degrades. Furthermore, the sensitivity of cancellation-phase to changes in lower primary-tone level decreases. This decrease forces one to greater primary frequency ratios causing reduced aural combination-tone amplitude and frequency, thus further adding difficulty to the task. As a result, no data could be obtained for subjects S1, S2, and S4 for $f_1 < 400$ Hz. Some data were obtained for S3 at the next lower 1/3-octave frequency, $f_1 = 315$ Hz; these data were essentially equivalent to the results for $f_1 = 400, 500, 630$ Hz.

7. Psychophysical measurements have also been made utilizing nonacoustic MEM activation, including: (e) the effects of voluntary MEM contraction on absolute thresholds (Smith, 1943; Reger, 1960; Reger et al, 1962) and on reducing temporary threshold shifts to high-intensity click stimuli (Fleer, 1962); and (f) the effects of electrically evoked MEM contraction (by the application of shocks to the external auditory meatus) on absolute thresholds (Pichler and Bornschein, 1957) and on suprathreshold loudness judgments (Gunn, 1973). The results from these studies are, however, difficult to relate to the effects produced by acoustic reflexes because non-acoustic MEM contractions are likely to be different from those associated with acoustic reflexes, either in terms of the strength of the muscle contractions which occur and/or the specific muscles involved (c.f., Wever and Lawrence, 1954, pp. 184-185; Jepsen, 1963; Lidén et al, 1970; Morgan and Dirks, 1975; Jeter, 1976).

8. In the 1968 paper, Borg noted an indication of "perfect regulation" or 1 dB/dB "limiting action" by the reflex at 500 Hz for a narrow range of elicitor intensities just above 20 dB re reflex threshold. However, no mention of this phenomenon appears in the 1974 paper, in which the measurements were made on a larger subject set (19 vs 4) and over a greater intensity range for the eliciting stimulus.

9. In Sec. III. C. (1) we indicated that our estimates of the reflex-threshold for admittance-change are likely to be "biased" high a few dB because we averaged the microphone output for 250 msec

beginning 380 (or 580) msec after elicitor onset. Our estimates of the reflex-threshold for transmission-change are also likely to be "biassed" high a few dB because a detectable transmission-change i.e. $\Delta\Theta_c \neq 0$, requires that the MEM contraction remain "on" and relatively stable for (at least) 3 to 5 sec after elicitor onset.

10. If "series" circuit is generalized to include an ideal transformer (of ratio k) between $U_d$ and $U_s$, then the statement that $\Delta|H| < \Delta|Y|$ implies shunting means that $U_d = kU_s + U_{shunt}$ with $U_{shunt} > 0$ for the resting and/or reflexed states, and this conclusion is valid if and only if k is unchanged by the MEM contraction.

11. Møller (1963) reached an equivalent conclusion from attempting to explain differences in $Y_d(f)$ between human and cat.

12. This represents an average of results from their Experiment II, Condition B, 102.2, and Condition C, 103.5, and Experiment III, Condition C, 102.3.

13. Details of these procedures are given in Nedzelnitsky (1974).

# FIGURE CAPTIONS

Fig. 1. Admittance-change, $\Delta Y_d(f)$, with the reflex-elicitor (contralateral-noise) level, $L_N$, as a parameter. Results shown for typical experiments on each subject, (a) S1 to (d) S4. The admittance-changes shown are relative to $Y_d(f;L_N=0)$ = averaged resting admittance for the subject on that day as derived from the sweeps with no contralateral noise ($L_N = 0$) which were carried out before, between, and after the "reflex" sweeps. Magnitude-changes, $\Delta|Y_d(f)|$ in dB $= 20\log_{10}|Y_d(f;L_N)/Y_d(f;L_N=0)|$, above; angle-changes, $\Delta\measuredangle Y_d(f)$ in degrees $= \measuredangle Y_d(f;L_N) - \measuredangle Y_d(f;L_N=0)$, below. Increasing elicitor levels are alternately coded with solid and dashed lines, and as follows: $L_N$ in dB SPL = 85 □, 95 △, 100 ○, 105 ▼, 110 ■, and 115 ▲. The standard deviations of the resting admittance for the test sessions, $\sigma[Y_d(f)]$ in dB and degrees, are also shown. The resting admittance averages and standard deviations were derived from 7 sweeps (with $L_N = 0$) for S1, 2 for S2, 4 for S3, and 4 for S4 for the sessions represented.

Fig. 2. Averaged values for "maximum admittance-change", MAX-$\Delta Y_d(f)$. Format as in Fig. 1. For each subject a measurement of MAX-$\Delta Y_d(f)$ was obtained as the $\Delta Y_d(f)$ for $L_N$ = 110 dB SPL. If in the same session a second measurement was made with $L_N$ = 110 or 115 dB SPL, these results were averaged. Measurements for 3 or 4 sessions (separated by 2 to 10 weeks each) were then averaged for each subject. These averaged admittance-changes, MAX-$\Delta Y_d(f)$, along

with the intersession standard deviations of the changes, $\sigma[\text{MAX-}\Delta Y_d(f)]$ in dB and degrees, are shown for each subject. The heavy line for $\bar{S}$ indicates an average MAX-$\Delta Y_d(f)$ (across the 4 subjects) obtained after shifting the individual data horizontally so that the transitions to positive magnitude-change all occur at 870 Hz, the geometric-mean-transition frequency.

Fig. 3. Average input admittance, $Y_d(f)$, with and without the "maximum admittance-change", MAX-$\Delta Y_d(f)$, produced by the acoustic reflex. The "RESTING" admittance function is the averaged data (for the 4 subjects) from Fig. 13 of Part 1. For the "MAX-$\Delta Y_d(f)$" function, the averaged MAX-$\Delta Y_d(f)$ from Fig. 2 (i.e. the heavy line for $\bar{S}$) was added, in dB and degrees, to the RESTING admittance function. Admittance magnitude (above): $|Y_d(f)|$ in dB re 1 cgs mS = $20\log_{10}[|Y_d(f)|/1 \text{ mS}]$, where 1 cgs acoustic mho = 1 $\text{cm}^5\text{dyne}^{-1}\text{sec}^{-1}$, and mS = millisiemen = mmho. Admittance angle (below): $\measuredangle Y_d(f)$ in degrees.

Fig. 4. Eardrum admittance components of conductance, $G_d(f)$ above, and susceptance, $B_d(f)$ below, for the average RESTING and MAX-$\Delta Y_d(f)$ conditions. Computed from the data in Fig. 3.

Fig. 5. Eardrum impedance components of resistance, $R_d(f)$ above, and reactance, $X_d(f)$ below, for the average RESTING and MAX-$\Delta Y_d(f)$ conditions. Computed from the data in Fig. 3.

Fig. 6. Eardrum impedance-change for the MAX-$\Delta Y_d(f)$. Results for each subject and $\bar{S}$. Resistance-change, $\Delta R_d(f)$ above; reactance-change, $\Delta X_d(f)$ below. Impedance-changes computed by using the $\Delta Y_d(f)$ data in Fig. 2 along with the resting $Y_d(f)$ measures for each subject from Fig. 14 of Part 1, converting the input admittance with and without MAX-$\Delta Y_d(f)$ to the impedance $Z_d(f) = R_d(f) + jX_d(f)$, and then subtracting each subject's (and $\bar{S}$'s) resting impedance from the reflexed impedance.

Fig. 7. Influence of the estimate of the residual earcanal volume, $V_C$, on eardrum admittance-change. Using the $Y_d(f)$ functions in Fig. 3 for both the RESTING and MAX-$\Delta Y_d(f)$ conditions, new values for $Y_d(f)$ were computed (using Eq. 7 of Part 1) corresponding to changes in $V_C$ estimation of -0.13 cc (▼), +0.09 cc (▲), and +0.19 cc (◆). The effects of these different $V_C$-estimates along with our conventionally processed results (●, replotted for $\bar{S}$ from Figs. 2 and 6) are shown: (a) in terms of the admittance magnitude-change, $\Delta|Y_d(f)|$ in dB, above, and admittance angle-change, $\Delta\measuredangle Y_d(f)$ in degrees, below, and (b) in terms of the impedance components of resistance- and reactance-change $\Delta R_d(f)$ above, and $\Delta X_d(f)$ below, both in cgs $\Omega$. Without the reflex active, the admittance functions $|Y_d(f)|$ and $\measuredangle Y_d(f)$ and the impedance components $R_d(f)$ and $X_d(f)$ are shown in Fig. 11 of Part 1 for these same alterations in $V_C$ estimation.

Fig. 8. Block diagram of apparatus used for generation of (a) combination-tone-complex stimuli, and (b) reflex-eliciting stimuli. Broken lines indicate equipment located within the soundproof room and under control of the subject (- - -) or the experimenter (-•-•-).

Fig. 9. Cancellation-phase ($\Theta_c$) dependence upon lower primary-tone level ($L_1$) for three values of primary-tone frequency ratio ($f_2/f_1$), all with $f_1$ = 630 Hz. Also, for $f_2/f_1 = 1.22$, dependence of $\Theta_c$ upon primary-tone amplitude ratio ($L_2/L_1$). An absolute phase reference for $\Theta_c$ (Goldstein, 1967) was not determined; the measured $\Theta_c$'s for the different values of $f_2/f_1$ have been arbitrarily shifted (vertically) for graphical presentation.

Fig. 10. Effects of acoustic reflexes on middle-ear transmission. Data for subjects S1 (top) to S4 (bottom) obtained using the $2f_1-f_2$ cancellation-phase method with $f_1$ and $f_2/f_1$ as indicated. The dotted lines connect averages of the data (at each value of $L_N$). Number pairs at the right of each dashed corner refer to the fitted straight-line segments. The first number indicates the values of $L_N$, in dB SPL, at the (projected) intersection of the straight-line segments; the second number is the magnitude of the slope of the straight-line segment on the right, in dB/dB.

Fig. 11. Average results for the acoustic-reflex effects on middle-ear transmission. Average of curves fitted to the data for four subjects with elicitor level, $L_N$, as a parameter. Vertical bars indicate

maximum and minimum (fitted) values over the four subjects for the circled points.

Fig. 12. Transmission-change for the acoustic reflex with the maximum elicitor level, $L_N$ = 110 dB SPL. Individual-subject (fitted) data, S1 to S4, and averaged results, $\bar{S}$. $\Delta|H(f)| = 0$ for $f \geqslant 1.25$ kHz for S1 and S4, for $f \geqslant 1.6$ kHz for S3, and for $f \geqslant 2.0$ kHz for S2 and $\bar{S}$.

Fig. 13. Effect of static pressure in the earcanal on middle-ear transmission at 400 Hz for subject S3. The data were obtained using two psychophysical methods: (a) the $2f_1-f_2$ cancellation-phase method [$\Delta\Theta_c$, ●], and (b) a pure-tone monaural loudness-balance method [LB, □]. The measurements were accomplished making use of the apparatus designed for our eardrum admittance studies (see Sec. II. B of Part 1). The electroacoustic-probe-assembly (EPA) was sealed into the earcanal and the acoustic stimuli for the tests were delivered to the subject's ear using the volume velocity source of the EPA. The static pressure system of the EPA was used to manipulate the earcanal static pressure, $P_{DC}$, between ±40 cm $H_2O$ re ambient atmospheric pressure. The $\Delta\Theta_c$ method was used as in the acoustic-reflex tests. For the $L_B$ method, a single tone of frequency $f_1$ and adjustable level $L_1$ was presented to the subject's ear. Using a control valve to isolate the earcanal from the pressure reservoir of the pump system, a test value pressure was established in the pump-system

reservoir while the earcanal pressure, $P_{DC}$, was zero. The valve was then switched allowing the test pressure to be rapidly communicated to the subject's ear (i.e. within a few seconds due to flow resistance in the pressure tube). The subject then adjusted $L_1$ to match the loudness perceived just prior to establishing $P_{DC}$. A second match was then made by releasing $P_{DC}$ to zero and having the subject adjust $L_1$ to match the loudness perceived just prior to the pressure release. The two matching changes, i.e. the two $\Delta L_1$'s, were averaged. For both methods, however, a further correction to the "raw data" was needed to obtain middle-ear transmission-change. Specifically, as $P_{DC}$ changes the eardrum admittance changes, and hence, the sound pressure at the eardrum, $P_d$, also changes. The magnitudes of these changes, $\Delta|P_d|$, at 400 Hz were measured vs $P_{DC}$ using the microphone in the EPA, and these changes have been incorporated in the data shown in the figure. While the influence of the static pressure on $P_d$ is large (e.g., $\Delta|P_d| \approx +4$ dB as $P_{DC}$ changes from 0 to ±40 cm $H_2O$), the corrections have been applied equally to the data from both psychophysical methods; therefore, they have no effect on the <u>differences</u> in the estimates of transmission-change (at a given $P_{DC}$) for the two methods.

Fig. 14. (See figure.)

Fig. 15. Normalized MEM response vs the level of various single-tone contralateral eliciting stimuli. The frequencies in kHz of the eliciting stimuli are indicated alongside the data. The sound-pressures of the eliciting stimuli in dB re reflex thresholds are given on the abscissas. The ordinates indicate the observed reflex-response (as given by an index of admittance-change) normalized by the maximum observed reflex-response within each study, i.e. each panel presents estimates of $100\% \cdot M_R[P_{dL}(f)]/\text{MAX-}M_R$. Mean results are given from Djupesland, Flottorp, and Winther (1967, Fig. 4, N = 20), for the normal hearing subjects of Peterson and Lidén (1972, Fig. 7, N = 29, 45, 45, 52, and 42 for eliciting stimulus frequency increasing from 0.25 to 4 kHz), and for the normal hearing subjects of Beedle and Harford (1973, Fig. 4, N = 10, results averaged for the two test sessions).

Fig. 16. Comparison of the effects of a given level of stapedius contraction (or acoustic-reflex response) on the magnitudes of admittance-change, $\Delta|Y_d(f)|$, and transmission-change, $\Delta|H(f)|$, in three species. The results for human are our average data for $L_N$ = 110 dB SPL taken from Figs. 2 and 11. The data for rabbit are from Møller (1965, Fig. 15). The data for cat are from Møller [1965, Figs. 13 and 14, with the values for $\Delta|Y_d(f)|$ in Fig. 14 being averaged with those calculated from Fig. 13].

Fig. 17. Zwislocki (1962) model for the human middle-ear: (a) block diagram, (b) circuit configuration, and (c) element values. Input voltage, $P_d$, represents the sound pressure at the eardrum. Input current, $U_d$, represents the volume velocity of the eardrum.

Fig. 18. Peake and Guinan (1967) model for the cat middle-ear: (a) block diagram, (b) circuit configuration, and (c) element values. Input voltage, $V_1$, is analogous to the sound pressure at the eardrum. Input current, $I_1$, is analogous to one-half the displacement of the malleus near the umbo. Output current, $I_2$, is analogous to the velocity of the stapes.

Fig. 19. Predictions for the effects of increased stiffness provided by tensioning the stapedius muscle and/or reducing the annular-ligament compliance in the middle-ear models proposed by (a) Zwislocki (1962) for human, and (b) Peake and Guinan (1967) for cat. [The models are given in Figs. 16 and 17.] The effects are shown for transmission-change magnitude, $\Delta|H(f)|$ [upper], admittance-change magnitude, $\Delta|Y_d(f)|$ [(a) middle, and (b) upper], and admittance-change angle, $\Delta\measuredangle Y_d(f)$ [lower]. Comparative data are also included. The magnitude data, $\Delta|H(f)|$ and $\Delta|Y_d(f)|$, are repeated from Fig. 16. The data on $\Delta\measuredangle Y_d(f)$ for human are from Fig. 2 (for $\bar{S}$), and for cat were calculated from Møller (1965, Fig. 13). For the predictions the stapedius-muscle/annular-ligament compliance was presumed to be included in $C_c$ and $C_2$ respectively for the Zwislocki and Peake-Guinan models, and the

effects were calculated for successive reductions by two from the normal compliance values, i.e. $C_c = 60/2^n$ and $C_2 = 0.76/2^n$. For the Zwislocki model we used $Y_d = U_d/P_d$ and $H = U_c/P_d$, where $U_c$ is the current flowing through the cochleo-stapedial complex, i.e. through $R_cL_cC_c$. For the Peake-Guinan model we used $Y_d = I_1/V_1$ and $H = I_2/V_1$, and because Møller's data were obtained with the bulla open and the septum removed, we set $1/C_m = 0$ (short-circuiting the middle-ear cavities).

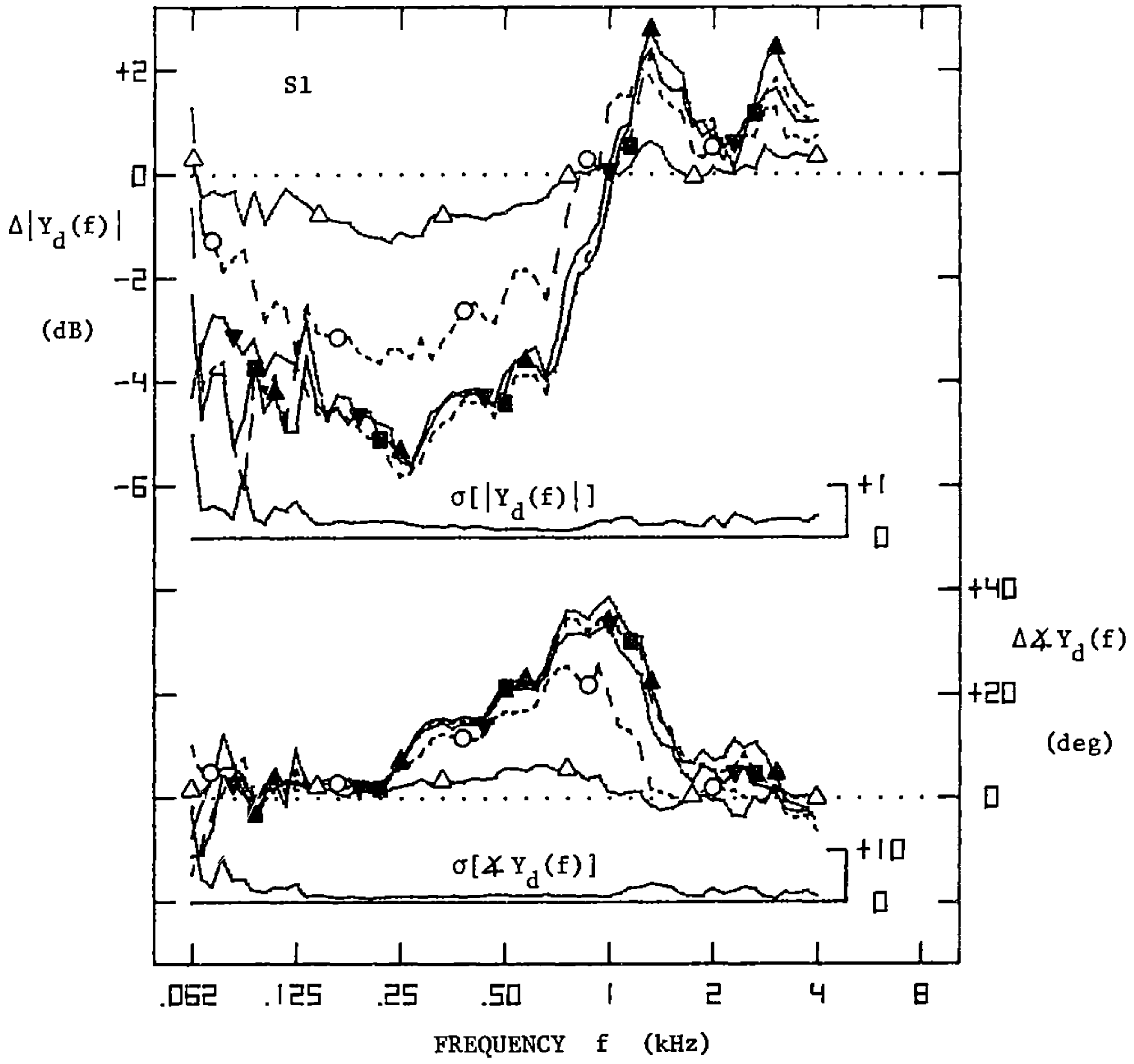

S1
+2
0
-2
-4
-6
Δ|Y_d(f)|
(dB)
σ[|Y_d(f)|]
+1
0
+40
+20
0
Δ∡Y_d(f)
(deg)
σ[∡Y_d(f)]
+10
0
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 1(a)

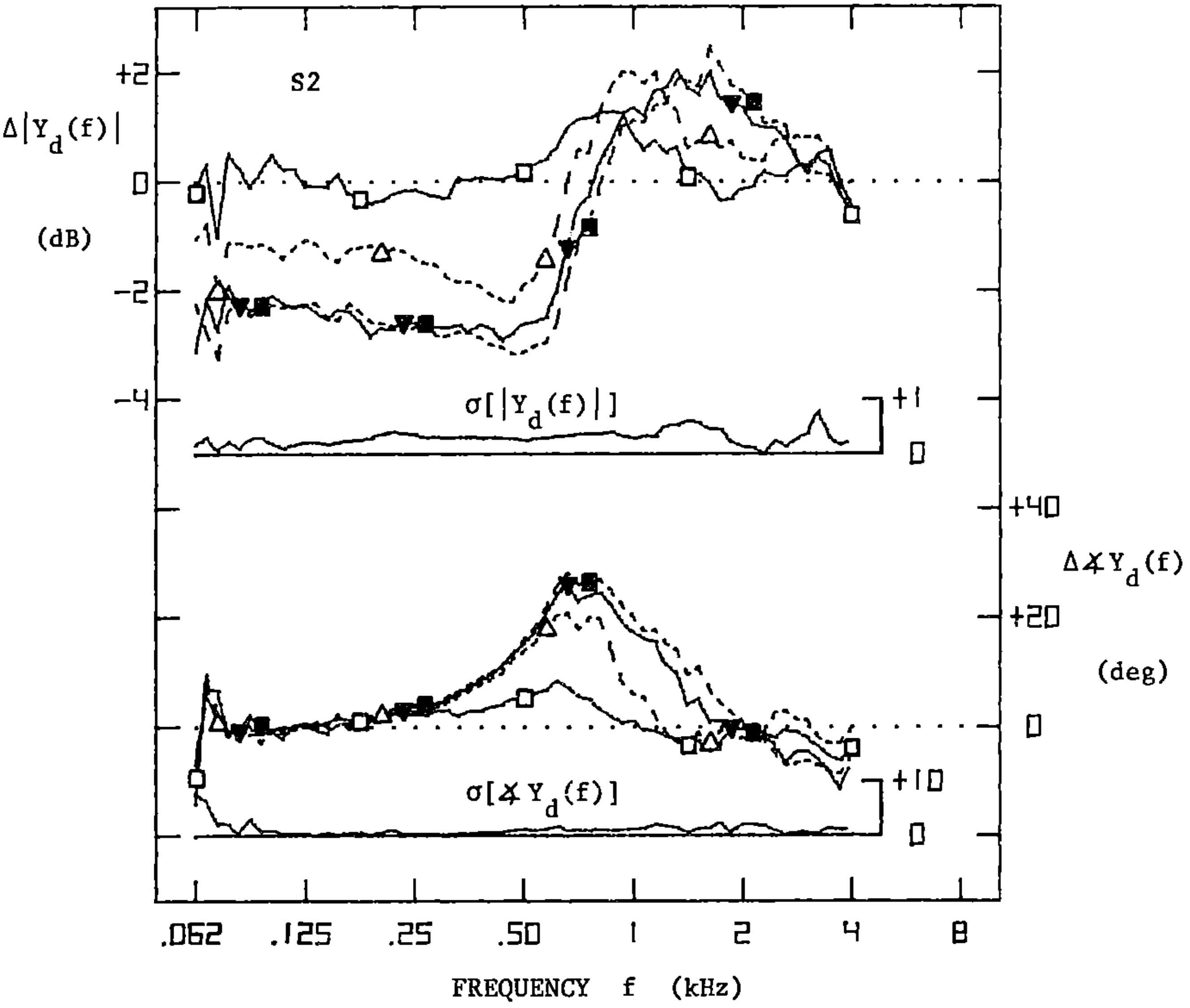

S2
Δ|Y_d(f)|
(dB)
+2
0
-2
-4
σ[|Y_d(f)|]
+1
0
+40
+20
0
Δ∡Y_d(f)
(deg)
σ[∡Y_d(f)]
+10
0
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 1 (b)

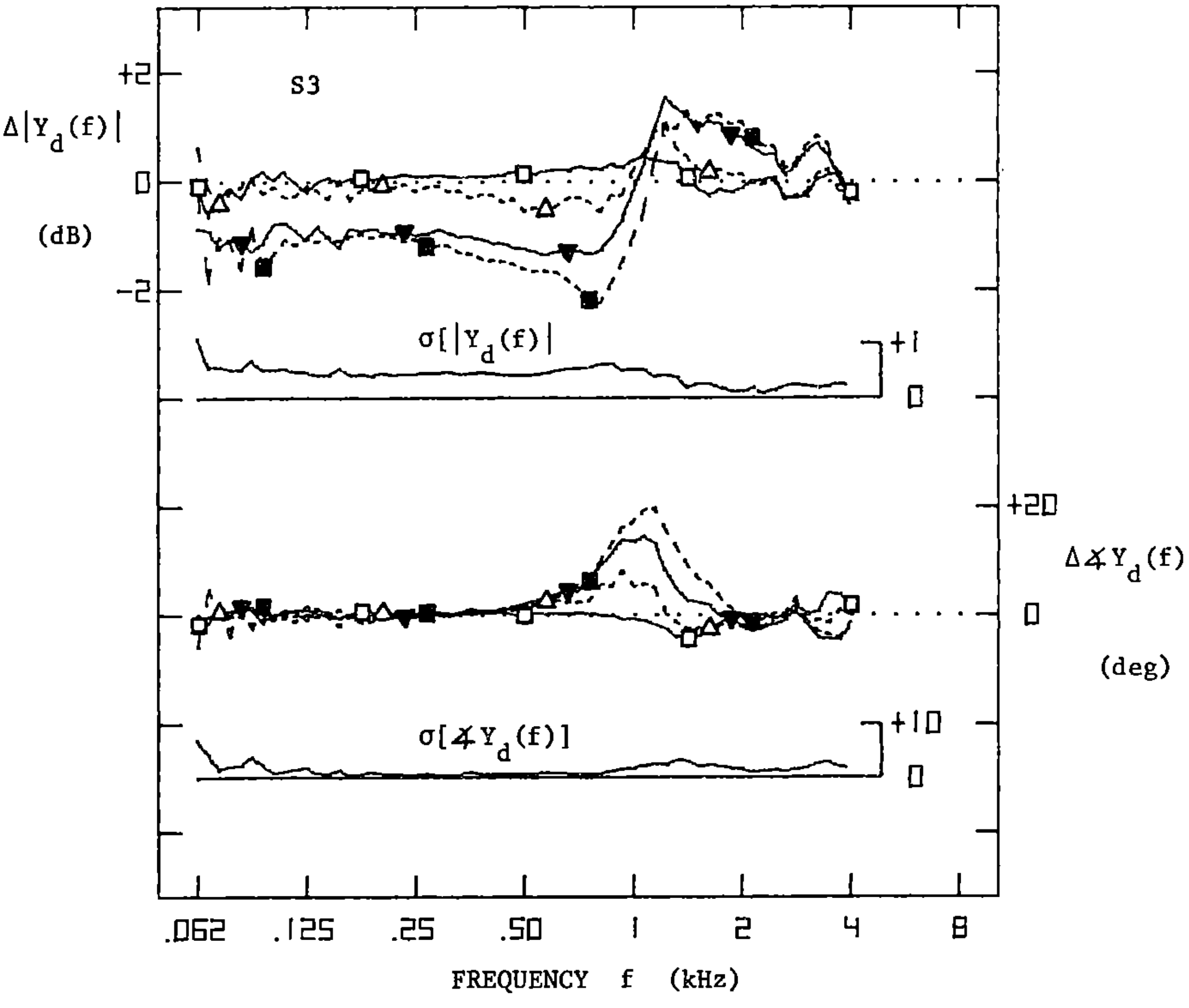

S3
+2
0
-2
Δ|Y_d(f)|
(dB)
σ[|Y_d(f)|
+1
0
+20
Δ∡Y_d(f)
0
(deg)
σ[∡Y_d(f)]
+10
0
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 1(c)

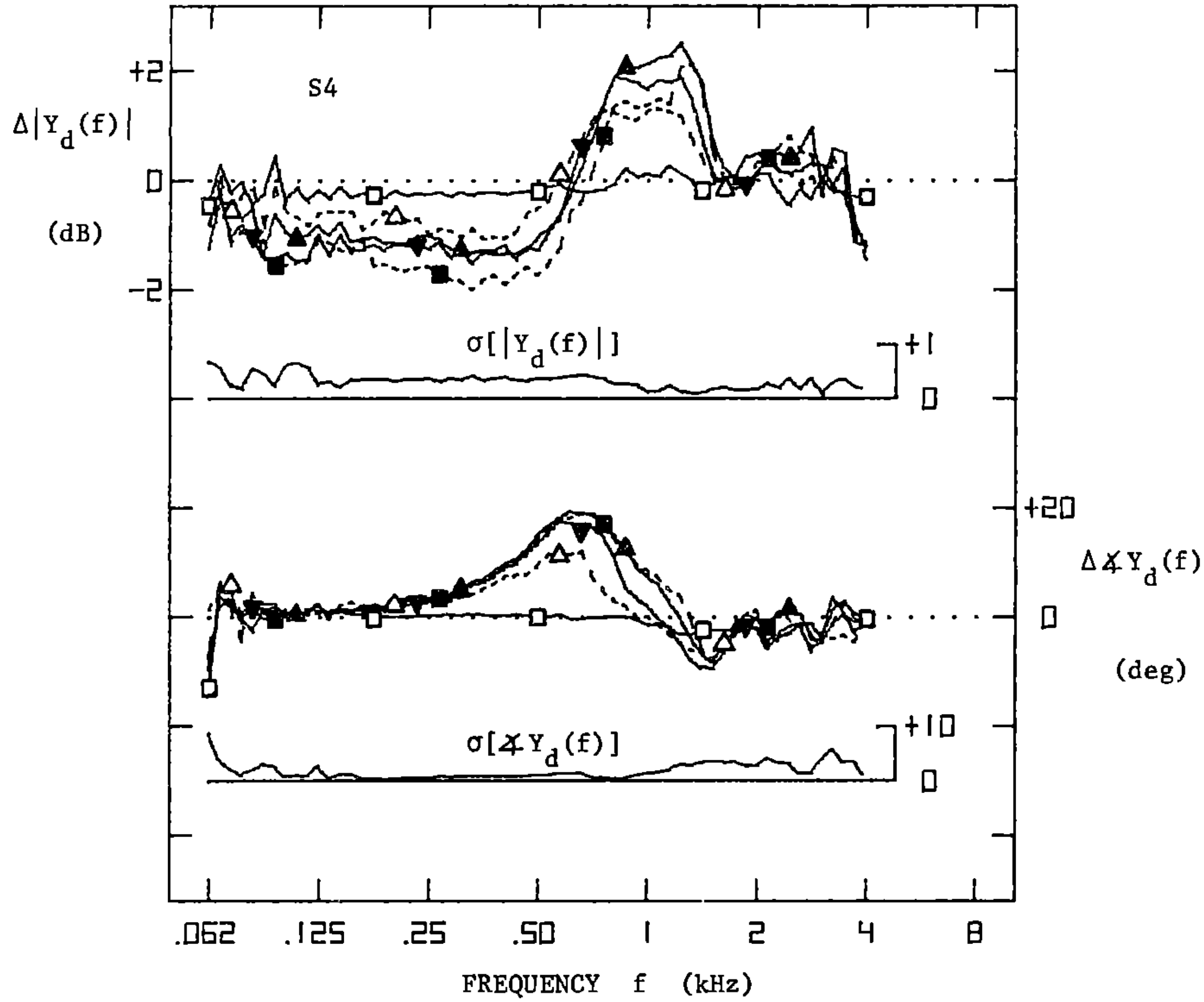

S4
+2
0
-2
Δ|Y_d(f)|
(dB)
σ[|Y_d(f)|]
+1
0
+20
0
Δ∡Y_d(f)
(deg)
σ[∡Y_d(f)]
+10
0
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 1(d)

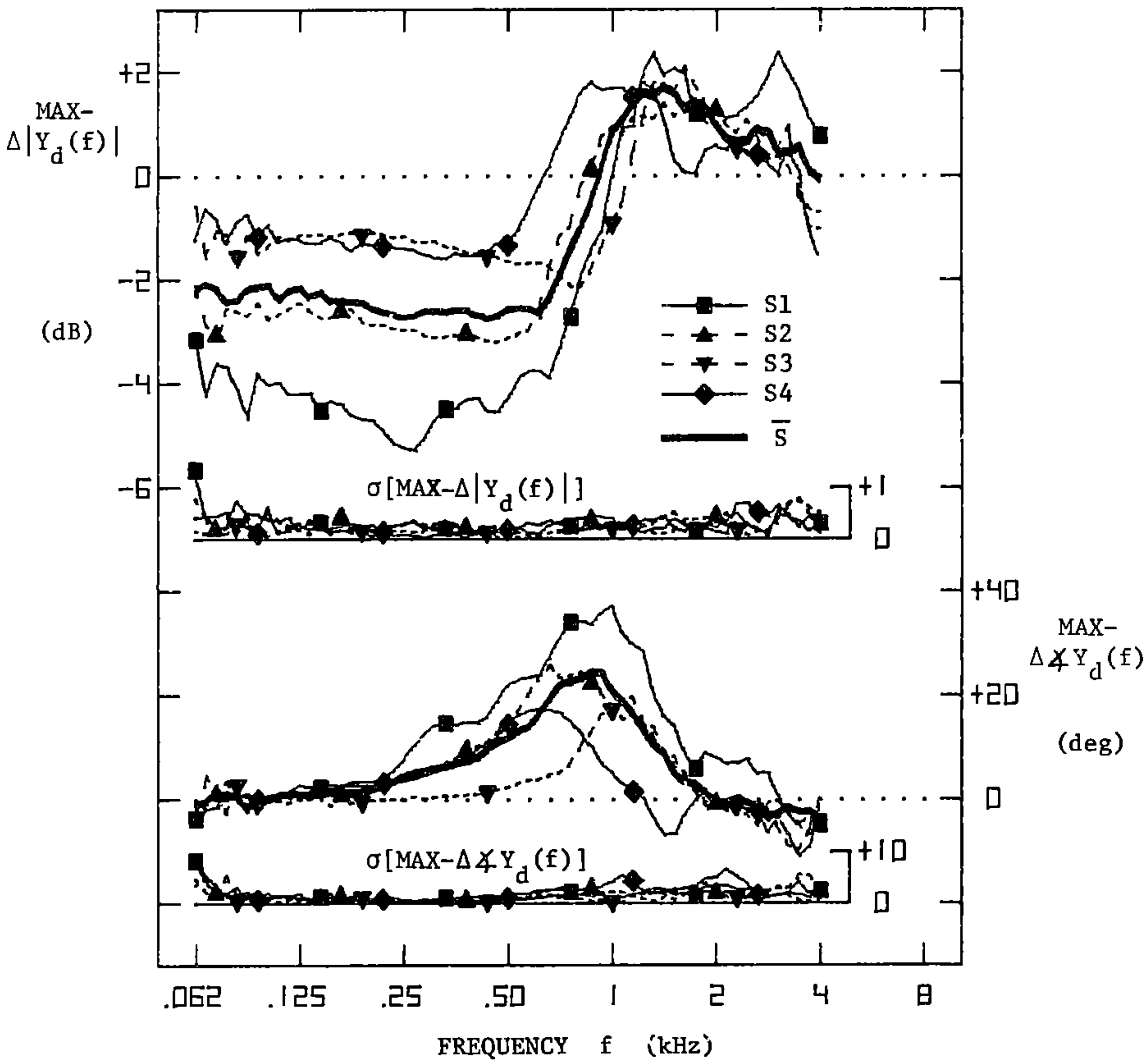
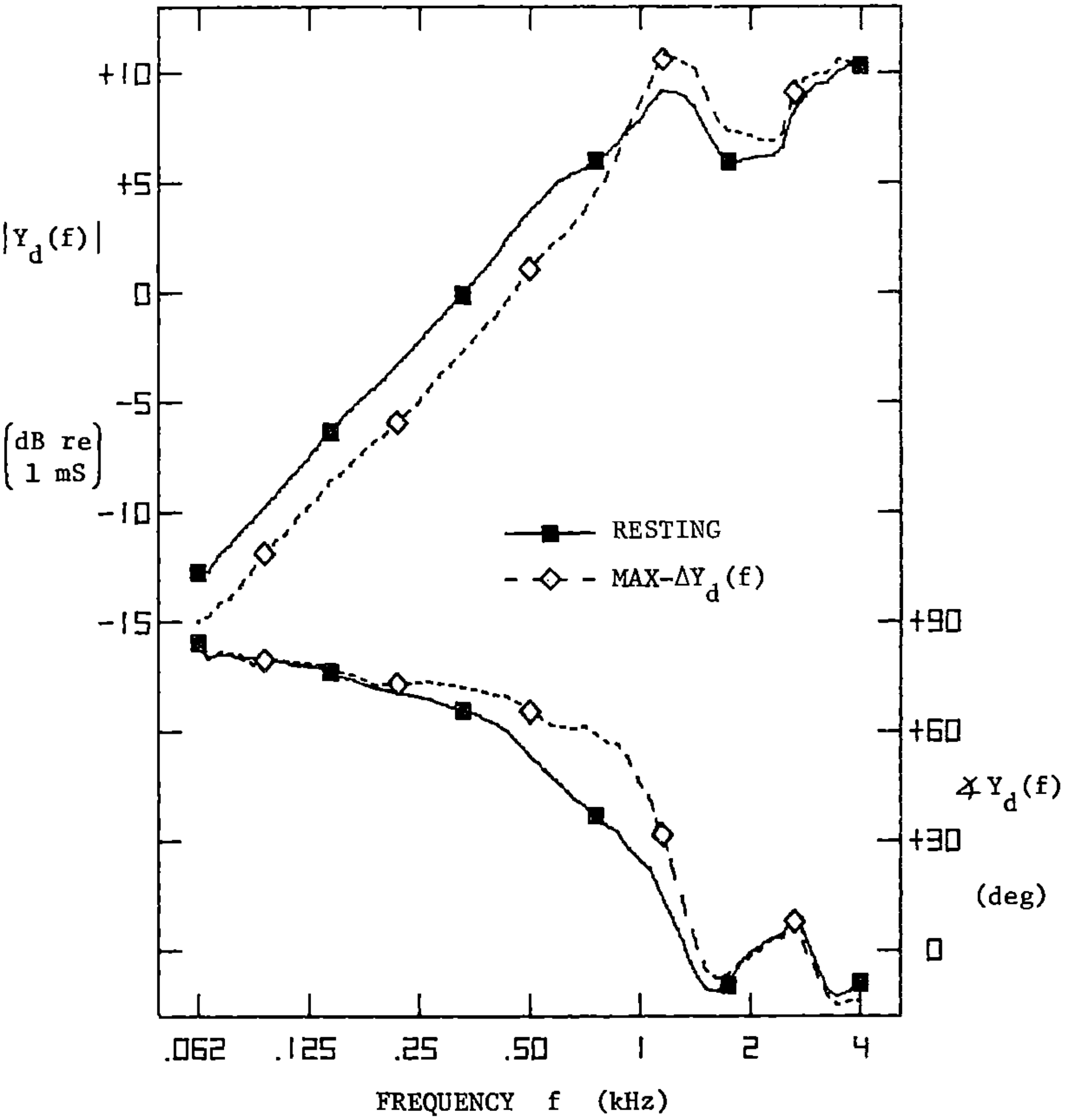

MAX-
$\Delta|Y_d(f)|$
(dB)
+2
0
-2
-4
-6
S1
S2
S3
S4
$\overline{S}$
$\sigma[\text{MAX-}\Delta|Y_d(f)|]$
+1
0
+40
+20
0
MAX-
$\Delta\measuredangle Y_d(f)$
(deg)
$\sigma[\text{MAX-}\Delta\measuredangle Y_d(f)]$
+10
0
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 2



+10
+5
$|Y_d(f)|$
0
-5
(dB re 1 mS)
-10
-15
+90
+60
+30
0
$\measuredangle Y_d(f)$
(deg)
RESTING
$\text{MAX-}\Delta Y_d(f)$
.062
.125
.25
.50
1
2
4
FREQUENCY f (kHz)

FIGURE 3

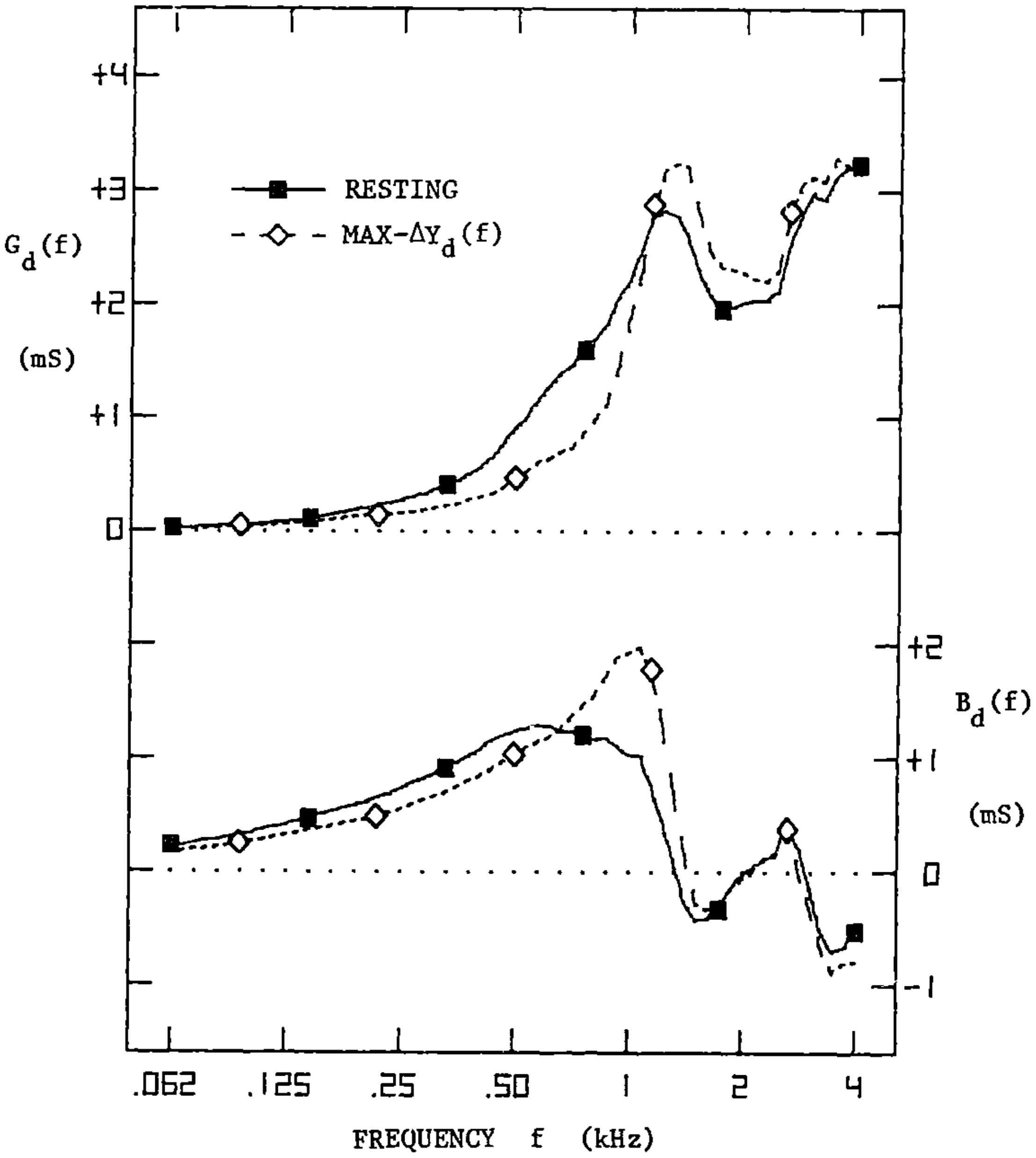
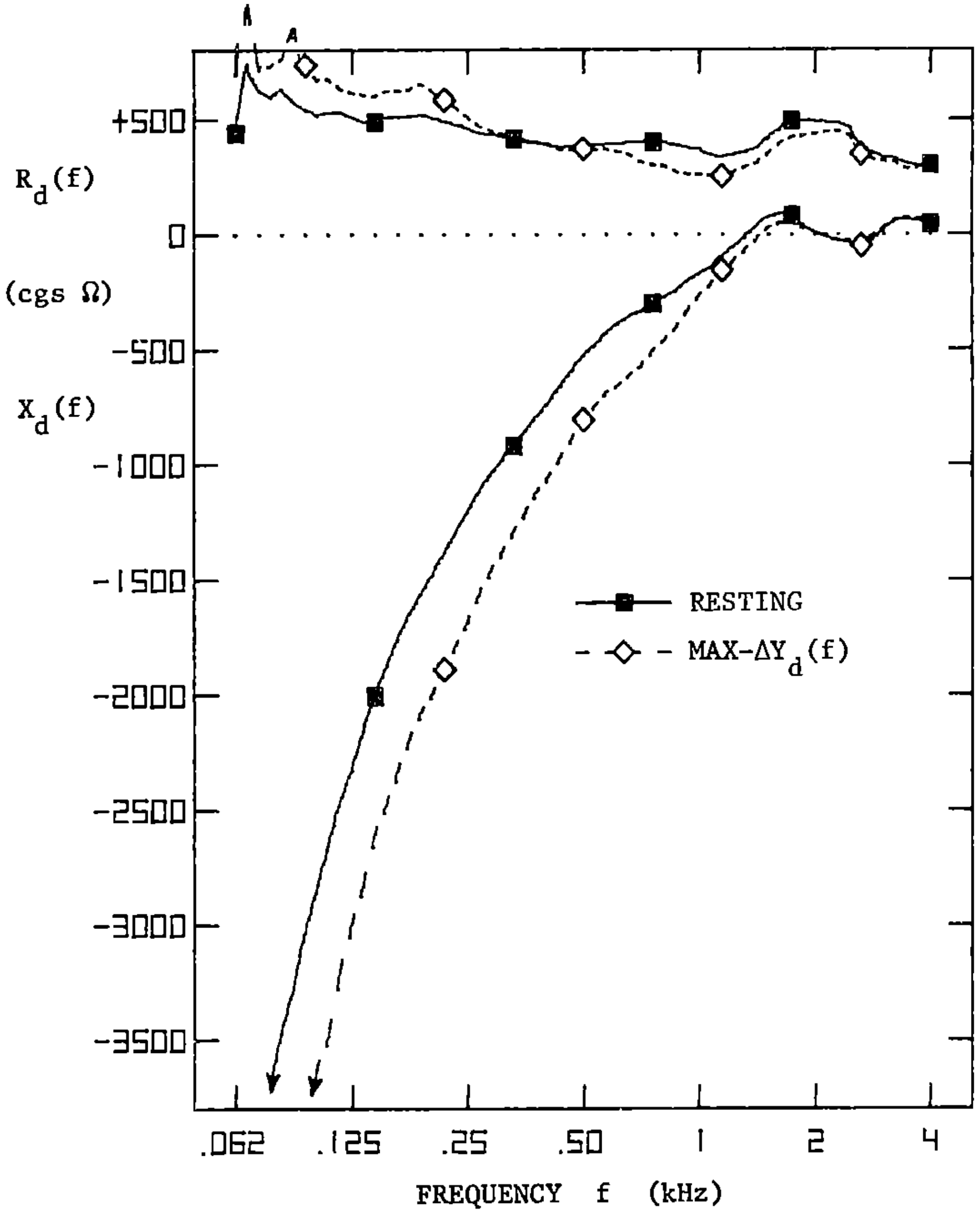

+4
+3
+2
+1
0
$G_d(f)$
(mS)
RESTING
MAX-$\Delta Y_d(f)$
+2
+1
0
-1
$B_d(f)$
(mS)
.062
.125
.25
.50
1
2
4
FREQUENCY f (kHz)

FIGURE 4



+500
$R_d(f)$
0
(cgs Ω)
-500
$X_d(f)$
-1000
-1500
-2000
-2500
-3000
-3500
.062
.125
.25
.50
1
2
4
FREQUENCY f (kHz)
■ RESTING
◇ MAX-$\Delta Y_d(f)$

FIGURE 5

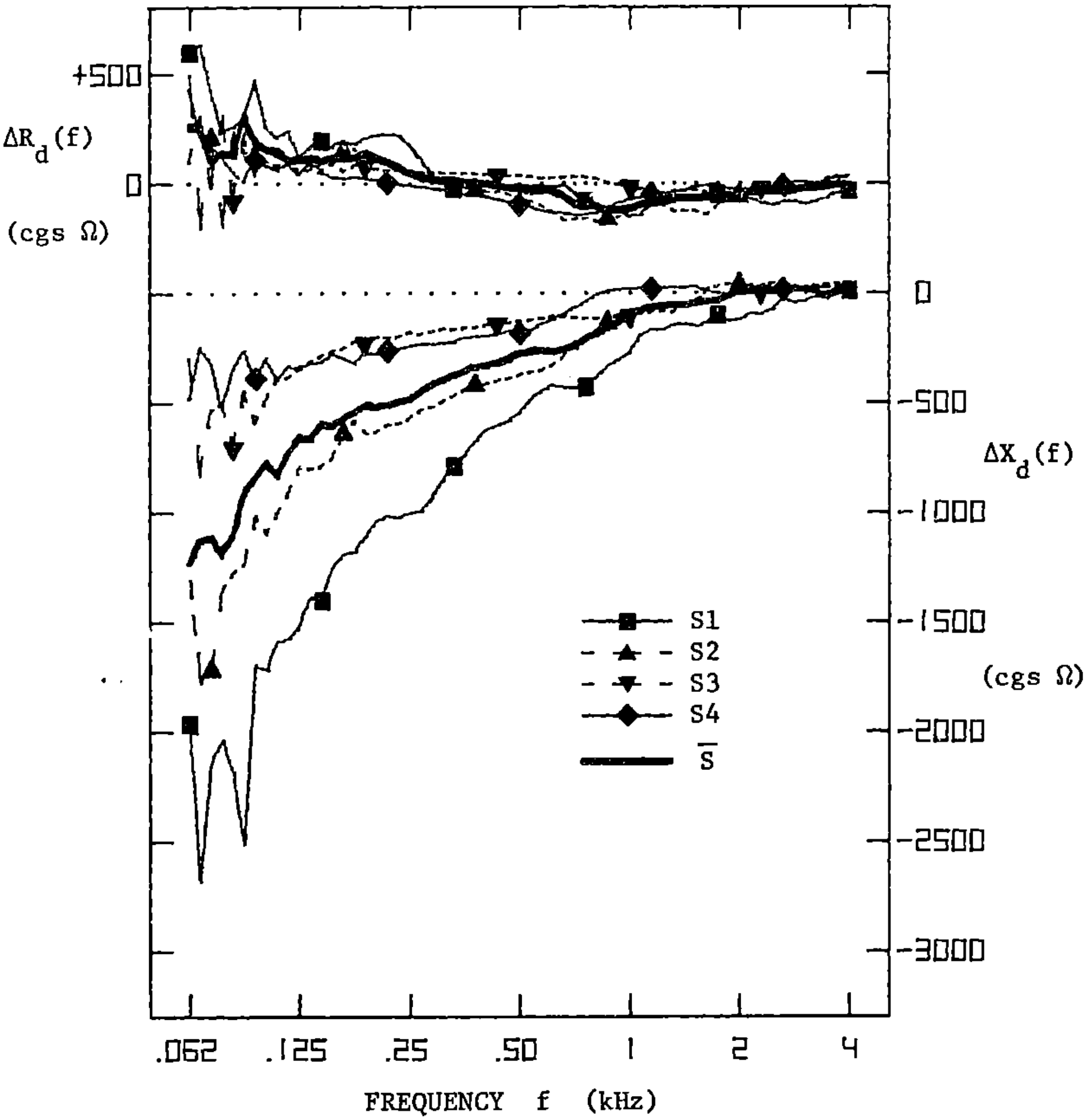

+500
$\Delta R_d(f)$
0
(cgs Ω)
0
-500
$\Delta X_d(f)$
-1000
-1500
(cgs Ω)
-2000
-2500
-3000
S1
S2
S3
S4
$\overline{S}$
.062
.125
.25
.50
1
2
4
FREQUENCY f (kHz)

FIGURE 6

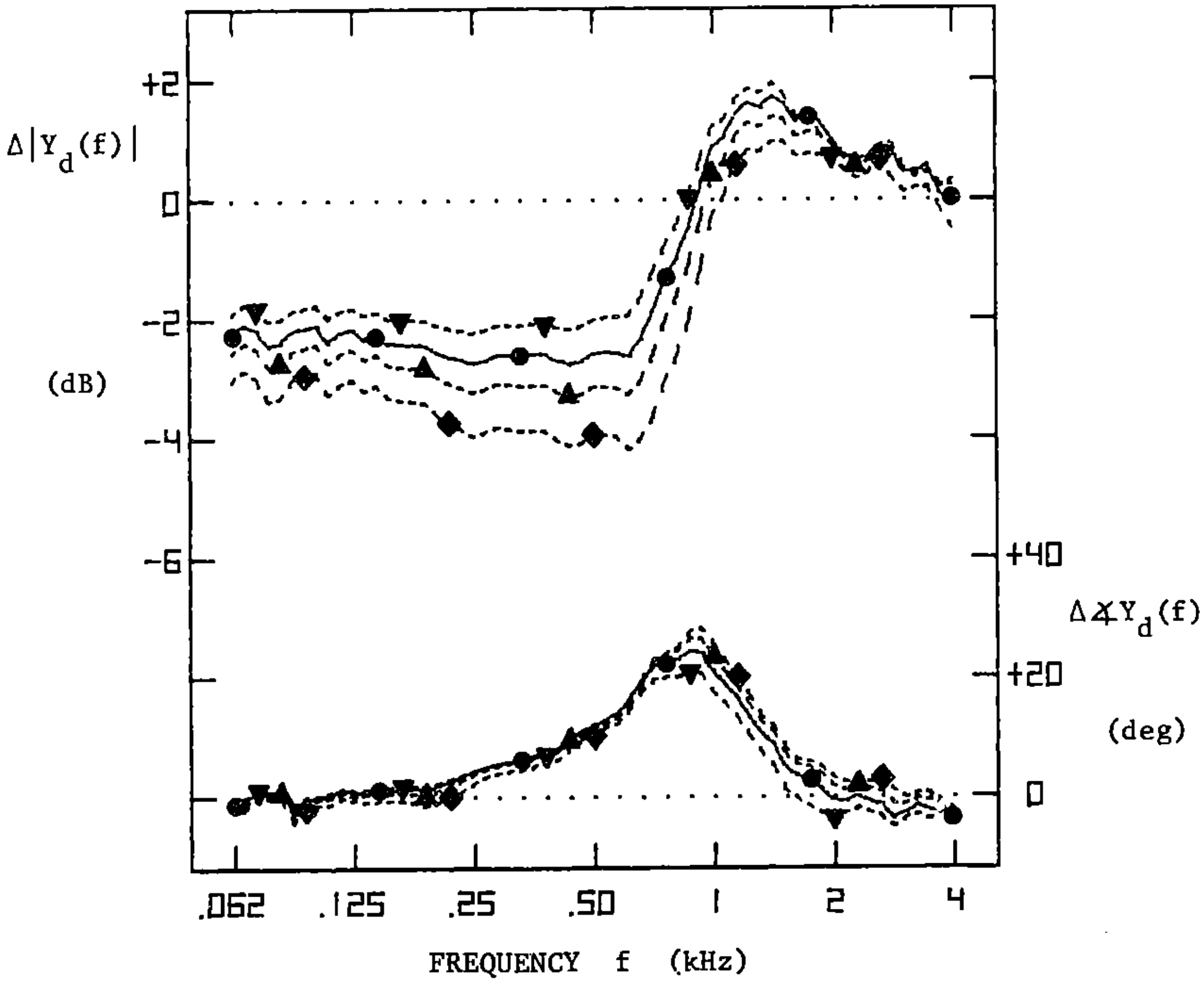


FIGURE 7 (a)

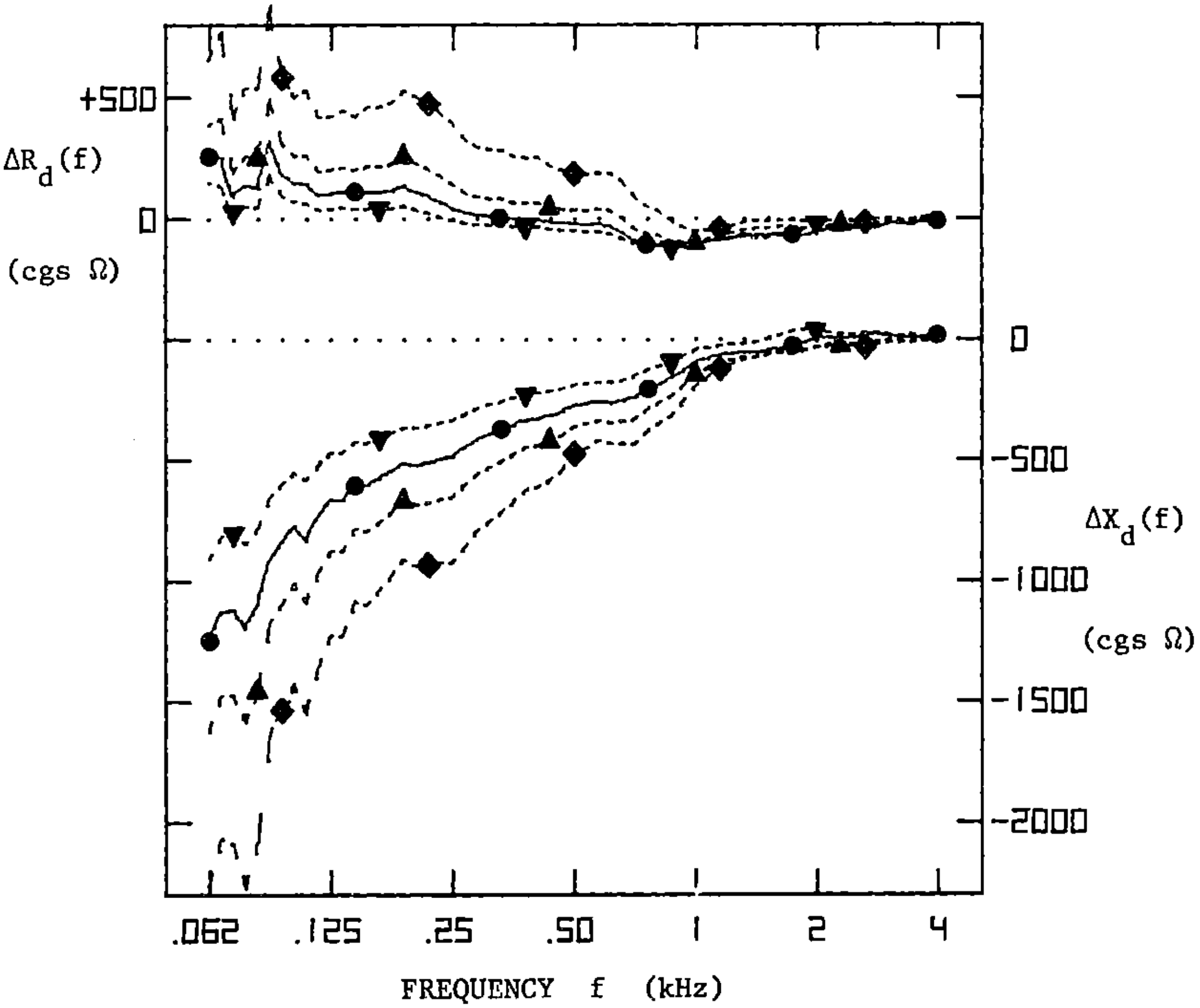

+500
ΔR_d(f)
0
(cgs Ω)
0
-500
ΔX_d(f)
-1000
(cgs Ω)
-1500
-2000
.062
.125
.25
.50
1
2
4
FREQUENCY f (kHz)

FIGURE 7(b)

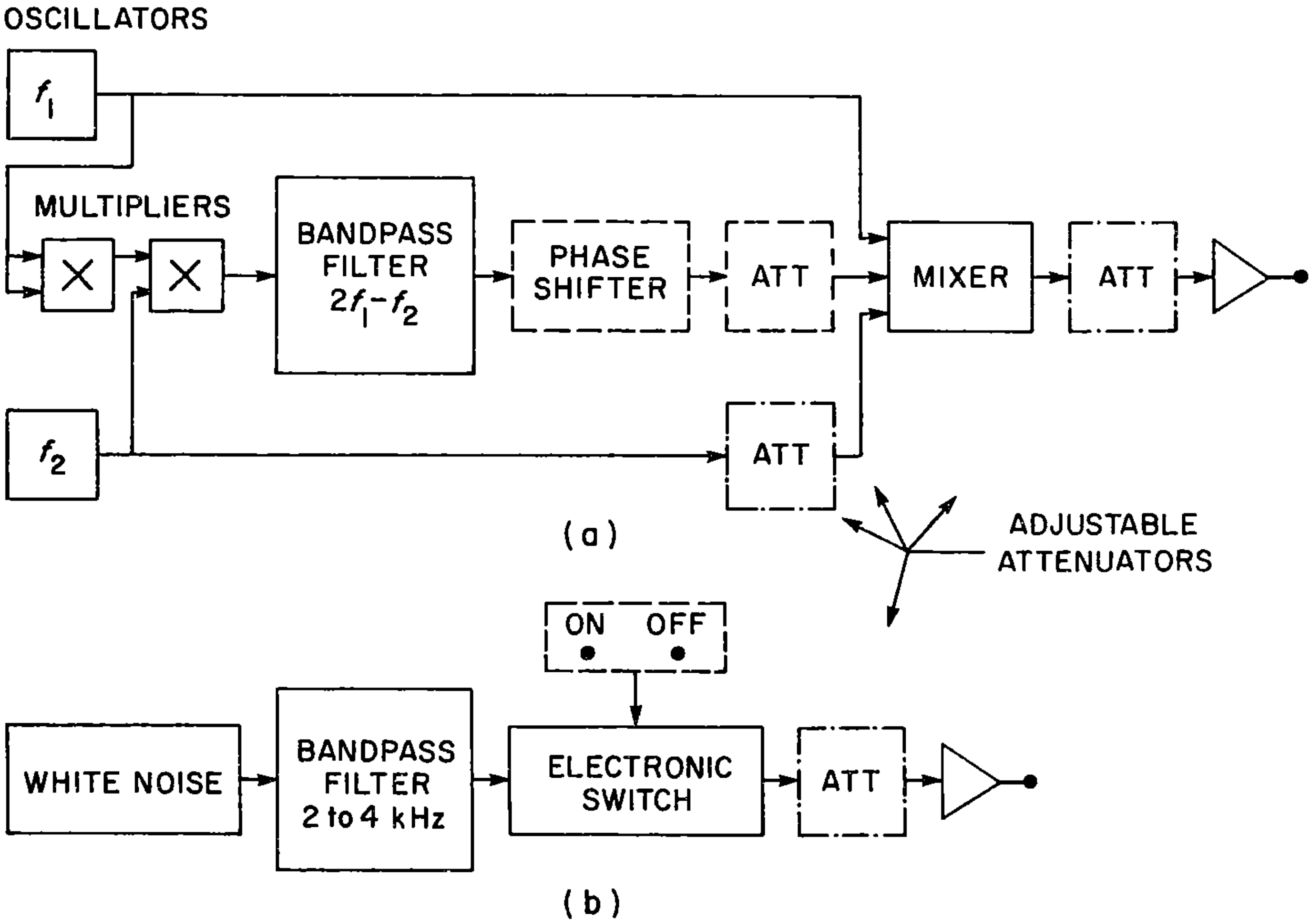

OSCILLATORS
$f_1$
MULTIPLIERS
×
×
BANDPASS FILTER $2f_1-f_2$
PHASE SHIFTER
ATT
MIXER
ATT
$f_2$
ATT
ADJUSTABLE ATTENUATORS
(a)
ON
OFF
WHITE NOISE
BANDPASS FILTER 2 to 4 kHz
ELECTRONIC SWITCH
ATT
(b)

FIGURE 8

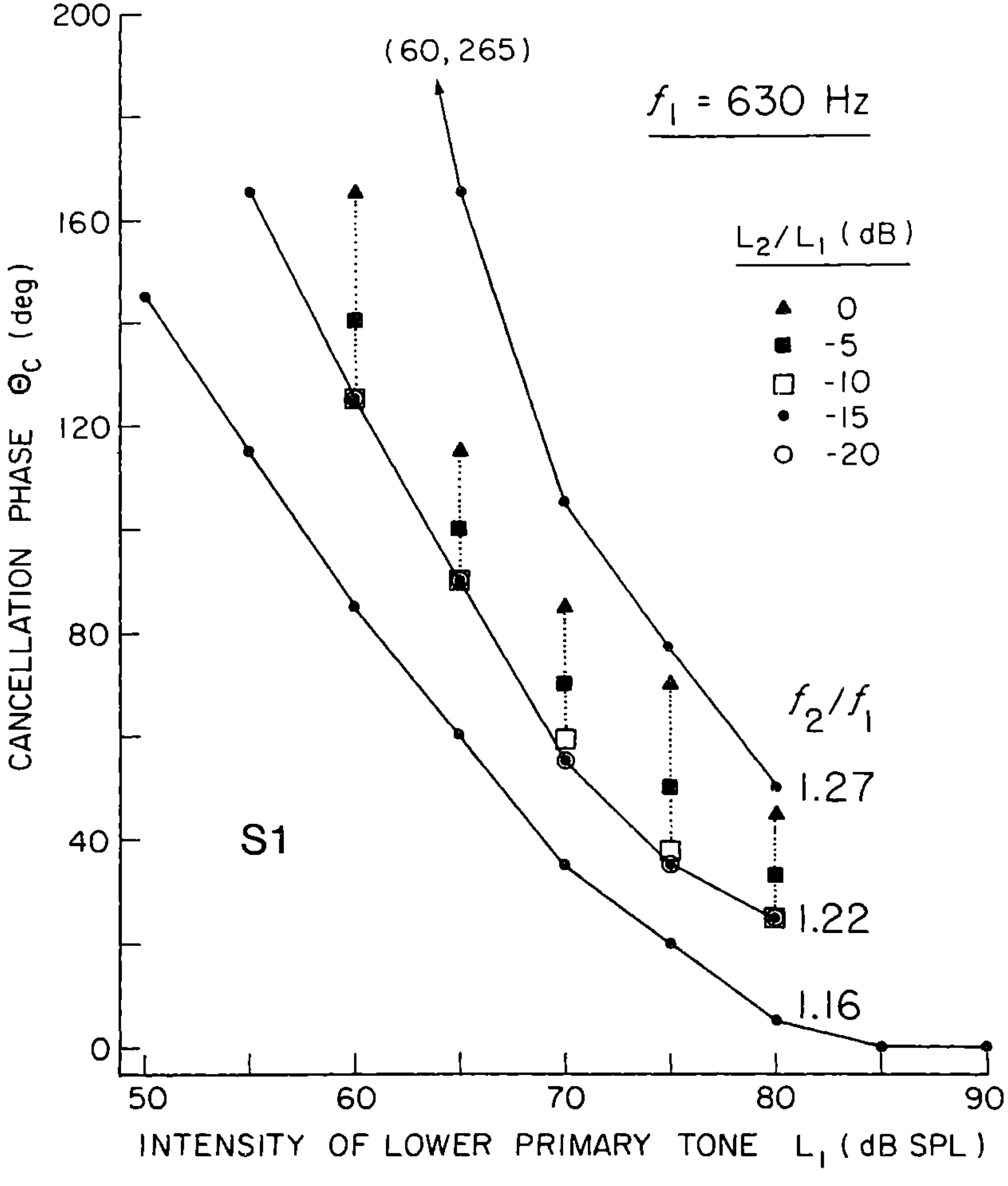

200
160
120
80
40
0
50
60
70
80
90
CANCELLATION PHASE $\Theta_C$ (deg)
INTENSITY OF LOWER PRIMARY TONE $L_1$ (dB SPL)
(60, 265)
$f_1$ = 630 Hz
$L_2/L_1$ (dB)
▲ 0
■ -5
□ -10
● -15
○ -20
$f_2/f_1$
1.27
1.22
1.16
S1

FIGURE 9

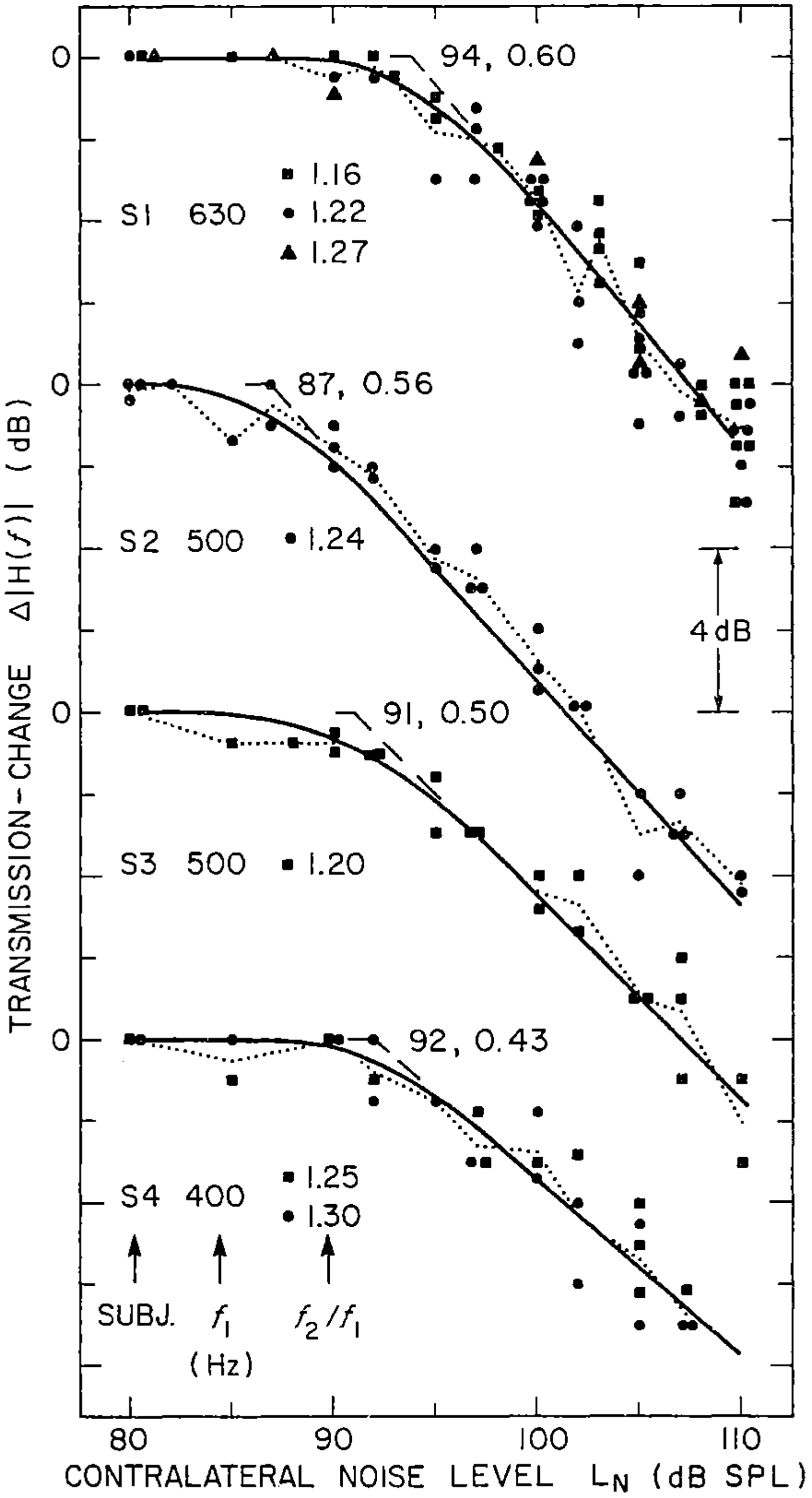

0
94, 0.60
■ 1.16
S1 630
● 1.22
▲ 1.27
0
87, 0.56
S2 500
● 1.24
4 dB
0
91, 0.50
S3 500
■ 1.20
0
92, 0.43
S4 400
■ 1.25
● 1.30
SUBJ.
$f_1$
(Hz)
$f_2/f_1$
TRANSMISSION - CHANGE $\Delta|H(f)|$ (dB)
80
90
100
110
CONTRALATERAL NOISE LEVEL $L_N$ (dB SPL)

FIGURE 10

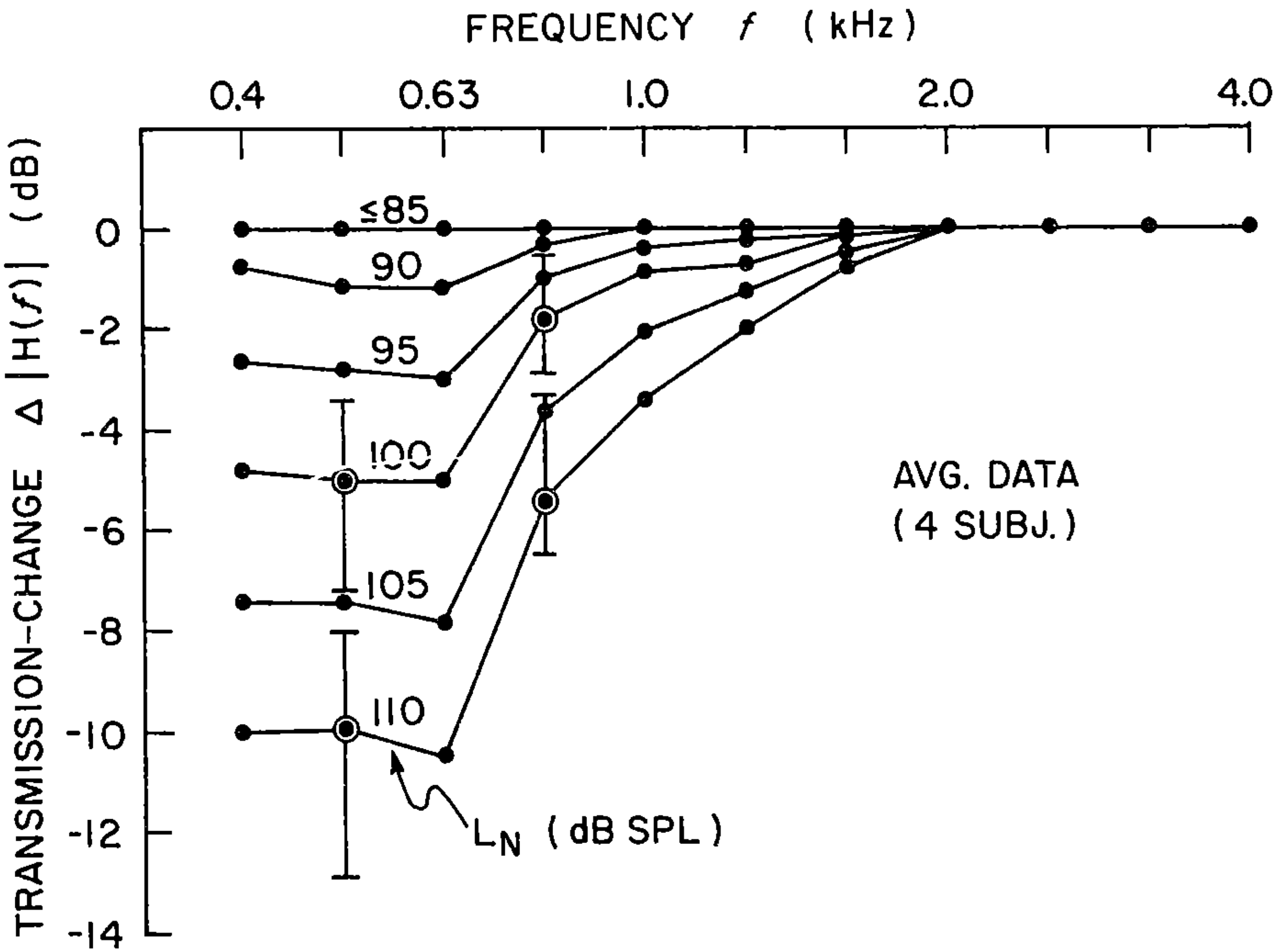

FREQUENCY f (kHz)
0.4
0.63
1.0
2.0
4.0
TRANSMISSION-CHANGE Δ |H(f)| (dB)
0
-2
-4
-6
-8
-10
-12
-14
≤85
90
95
100
105
110
L_N (dB SPL)
AVG. DATA
(4 SUBJ.)

FIGURE 11

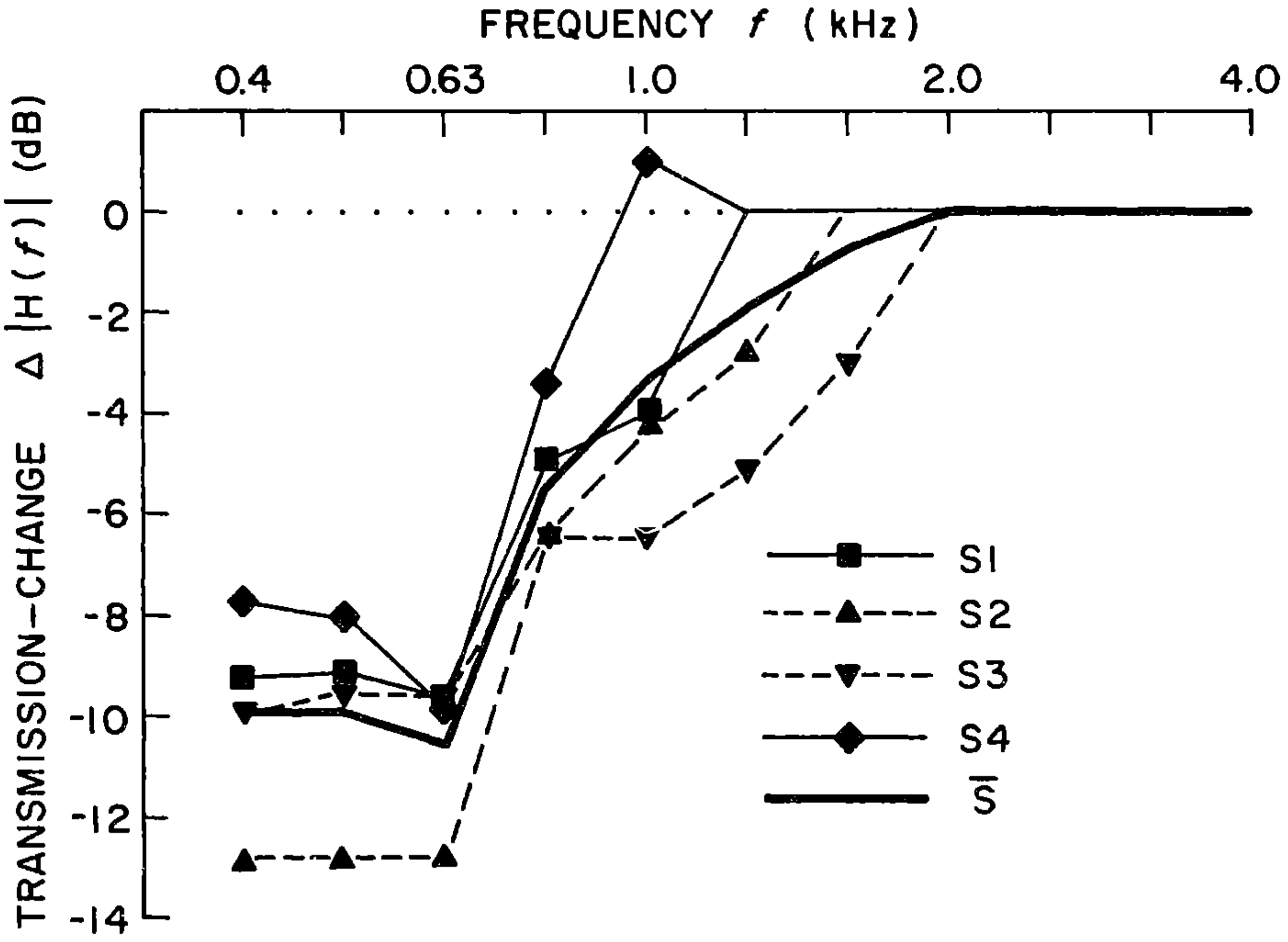

FREQUENCY $f$ (kHz)
0.4
0.63
1.0
2.0
4.0
TRANSMISSION-CHANGE $\Delta |H(f)|$ (dB)
0
-2
-4
-6
-8
-10
-12
-14
S1
S2
S3
S4
$\overline{S}$

FIGURE 12

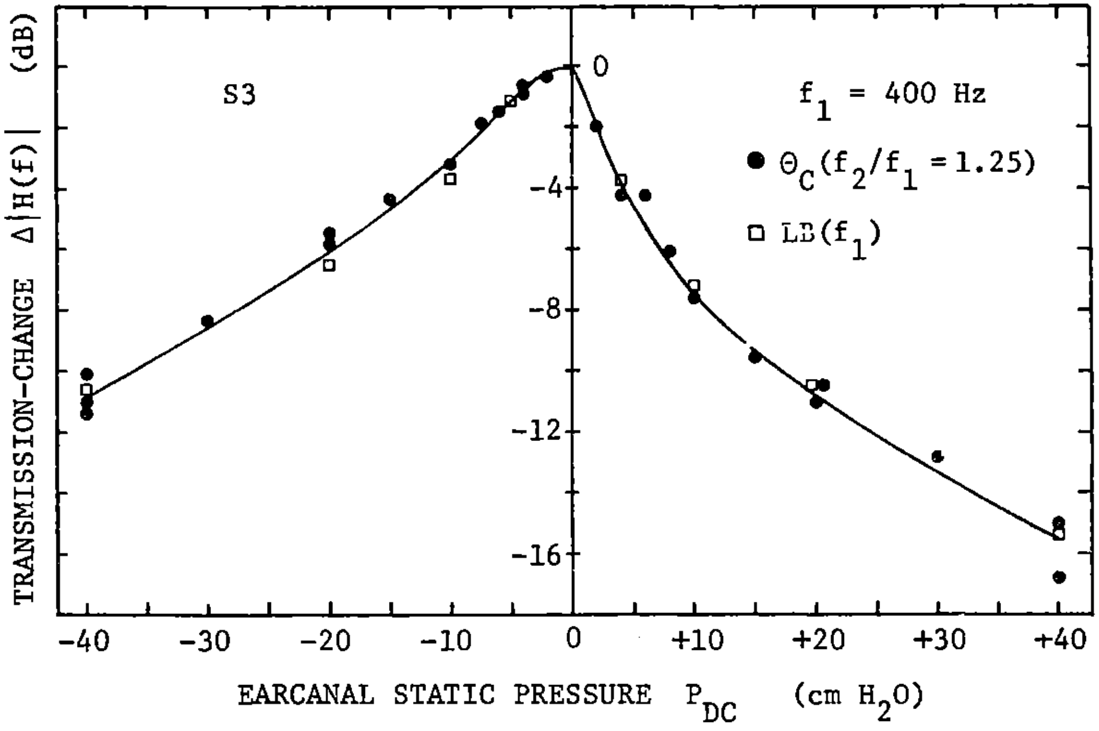

S3
f1 = 400 Hz
● ΘC(f2/f1 = 1.25)
□ LB(f1)
TRANSMISSION-CHANGE Δ|H(f)| (dB)
0
-4
-8
-12
-16
-40
-30
-20
-10
0
+10
+20
+30
+40
EARCANAL STATIC PRESSURE PDC (cm H2O)

FIGURE 13

Block diagram of the auditory system relevant for consideration of the effects of acoustic reflexes. Subscripts L and R denote left and right ears respectively. $P_d$ = sound pressure at the eardrum, $U_s$ = volume velocity of the stapes, $H = U_s/P_d$ = middle-ear transfer function, $\Delta Y_d$ = eardrum admittance-change, and M = signals sent to the MEM by the nervous system. Signals in M produce frequency dependent changes in admittance and transmission, i.e. $M \rightarrow \Delta Y_d(f)$ and $\Delta H(f)$. The stimuli used in measuring are assumed to be of sub-reflex eliciting level, i.e. they make no contribution to M; hence, the $\Delta Y_d$'s are indicated as "external" measurements made upon the middle-ears. For convenience, we assume that the transmission-change measurements are always performed on the right ear.

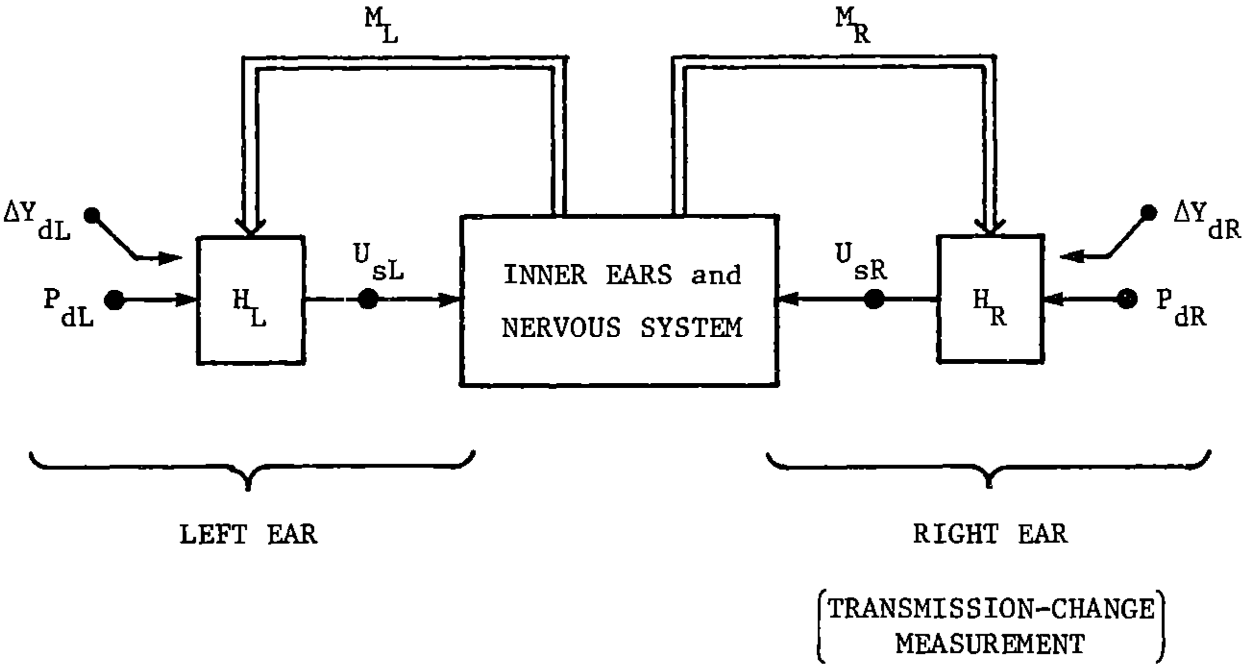


FIGURE 14

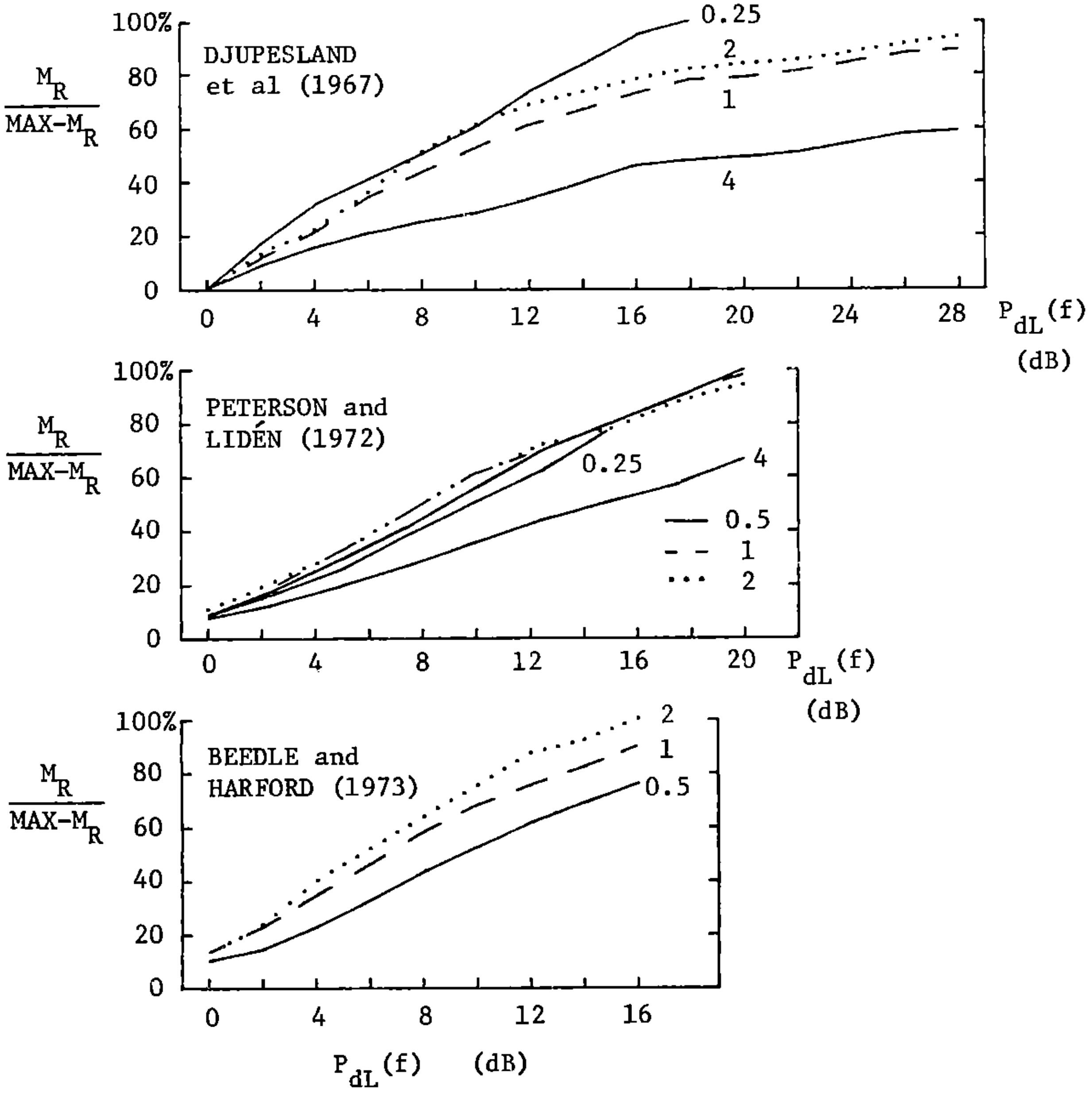
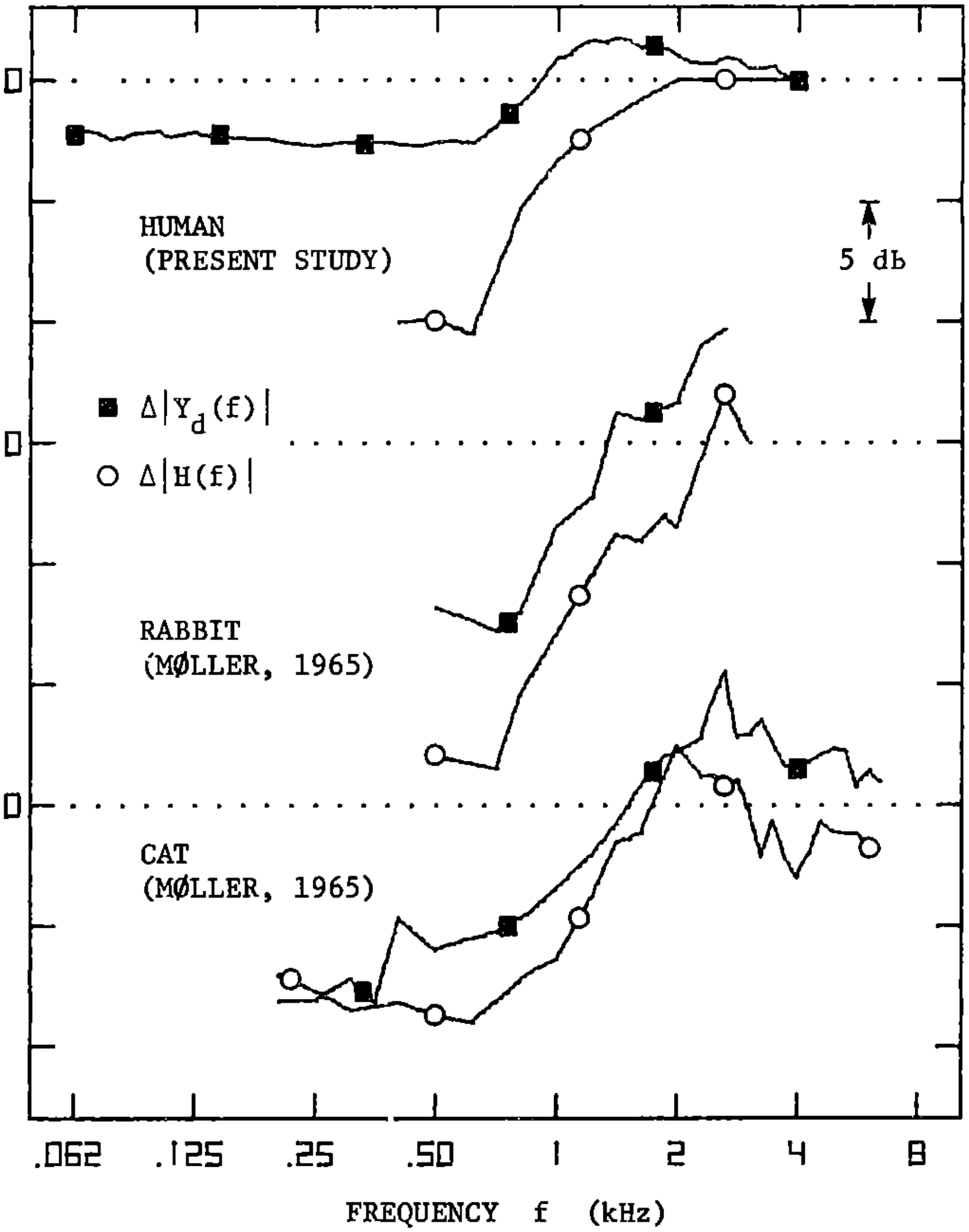
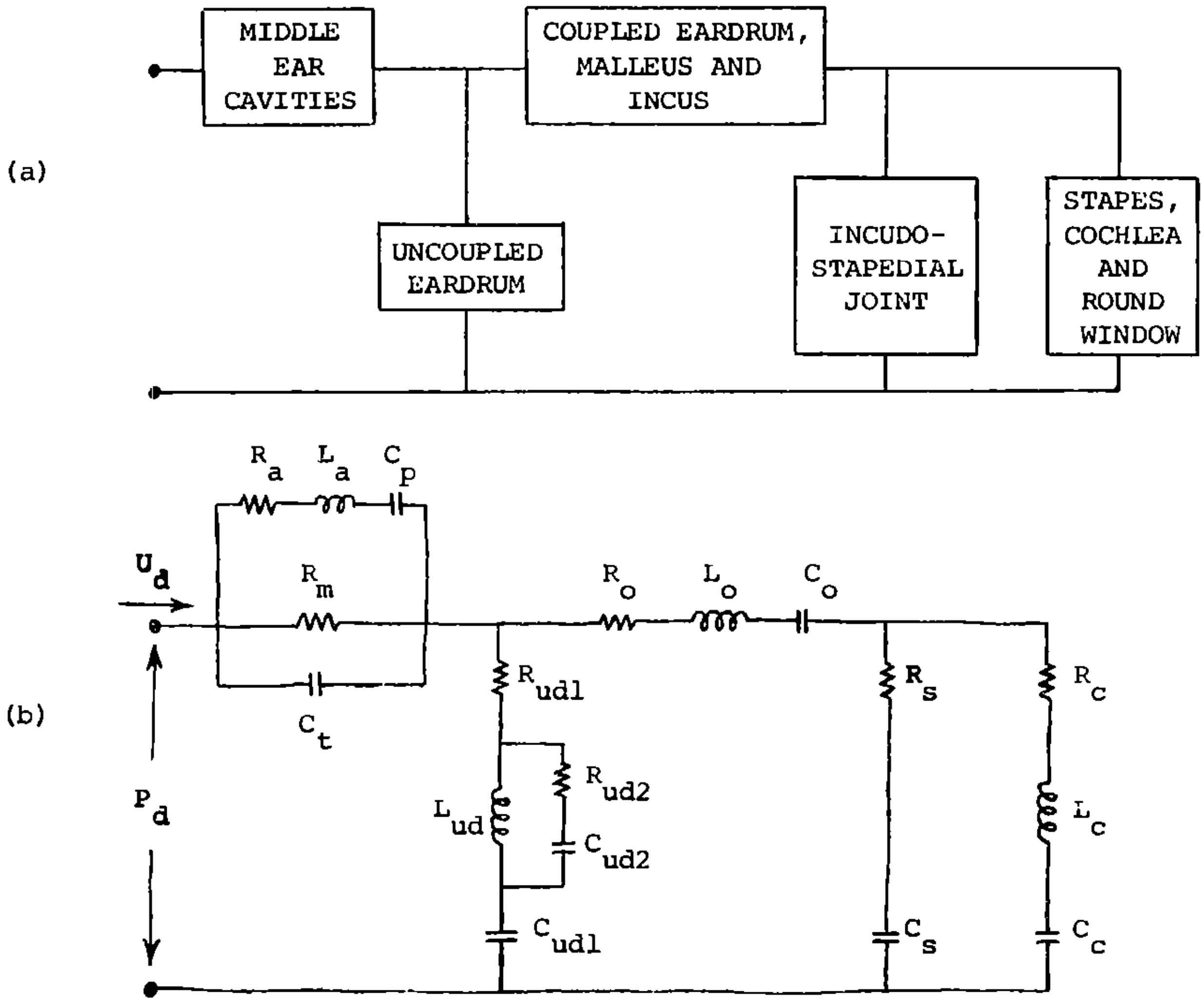

DJUPESLAND
et al (1967)
$M_R$ / MAX-$M_R$
100%
80
60
40
20
0
0
4
8
12
16
20
24
28
$P_{dL}(f)$ (dB)
0.25
2
1
4
PETERSON and
LIDÉN (1972)
$M_R$ / MAX-$M_R$
100%
80
60
40
20
0
0
4
8
12
16
20
$P_{dL}(f)$ (dB)
0.25
4
0.5
1
2
BEEDLE and
HARFORD (1973)
$M_R$ / MAX-$M_R$
100%
80
60
40
20
0
0
4
8
12
16
$P_{dL}(f)$ (dB)
2
1
0.5

FIGURE 15



0
0
0
HUMAN
(PRESENT STUDY)
5 db
■ Δ|Y_d(f)|
○ Δ|H(f)|
RABBIT
(MØLLER, 1965)
CAT
(MØLLER, 1965)
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 16



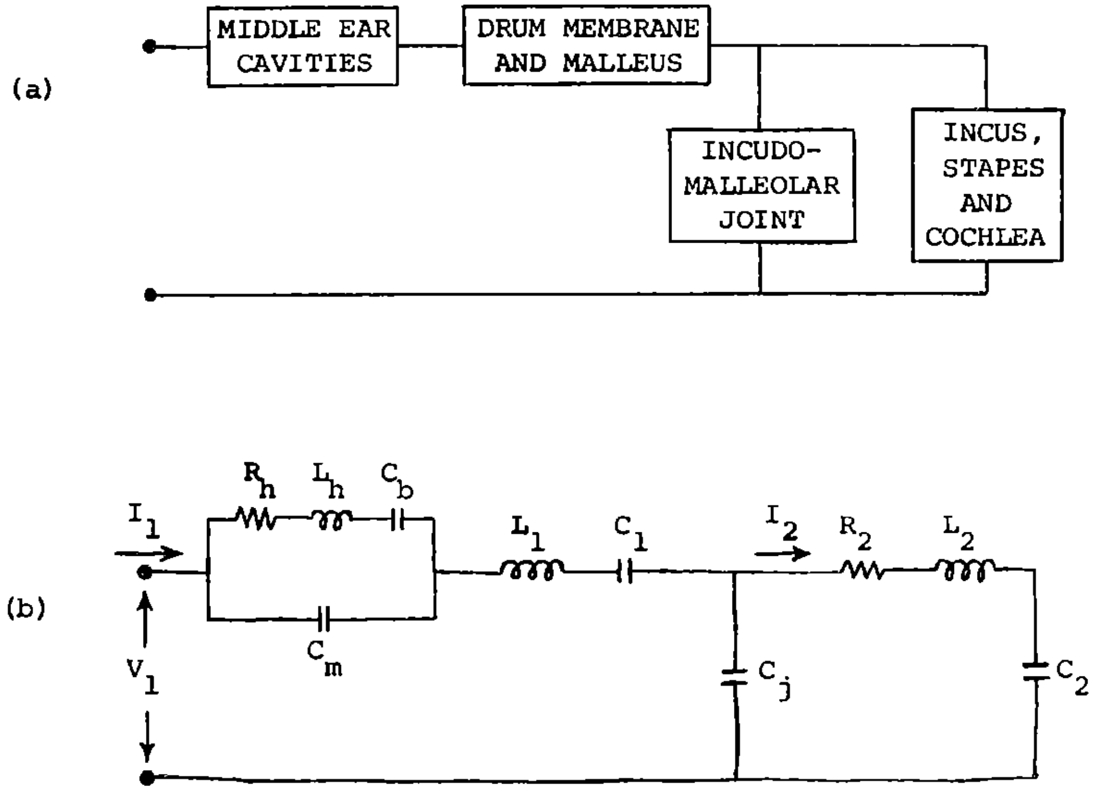
(c)

$R_a$ = 10 Ω
$R_m$ = 390
$R_{ud1}$ = 40
$R_{ud2}$ = 220
$R_o$ = 70
$R_s$ = 3000
$R_c$ = 600

$L_a$ = 14 mH
$L_{ud}$ = 15
$L_o$ = 40
$L_c$ = 20

$C_p$ = 5.1 uF
$C_t$ = 0.35
$C_{ud1}$ = 0.23
$C_{ud2}$ = 0.40
$C_o$ = 1.4
$C_s$ = 0.25
$C_c$ = 0.60

FIGURE 17



(c)

$R_h = 22\,\Omega$

$R_2 = 390$

$L_h = 14$ mH

$L_1 = 22$

$L_2 = 7.5$

$C_b = 0.30$ uF

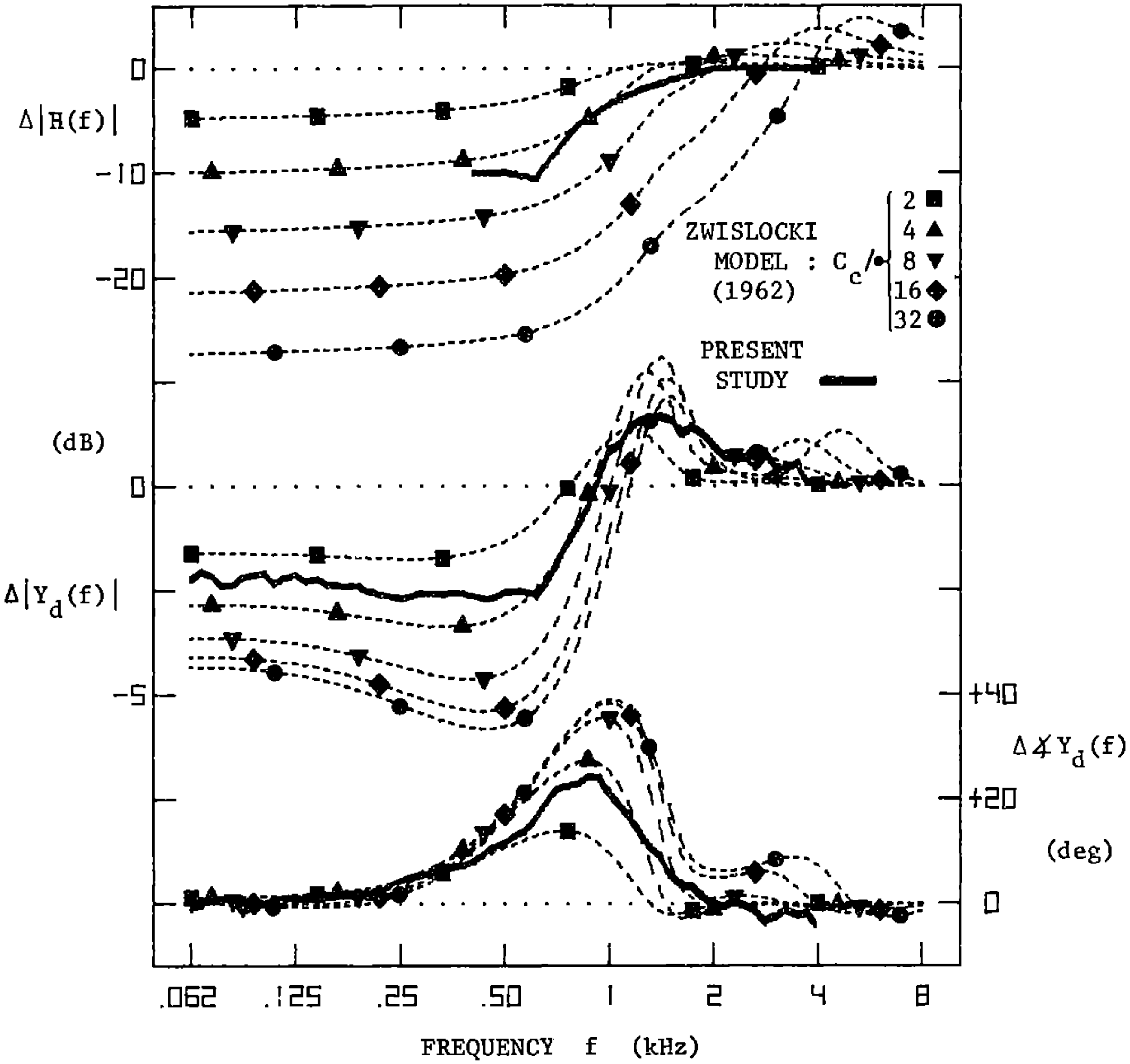
$C_m = 0.19$

$C_1 = 0.76$

$C_j = 0.042$

$C_2 = 0.76$

FIGURE 18



0
Δ|H(f)|
-10
-20
(dB)
0
Δ|Y_d(f)|
-5
+40
+20
0
Δ∡Y_d(f)
(deg)
ZWISLOCKI
MODEL : C_c/•
(1962)
2 ■
4 ▲
8 ▼
16 ◆
32 ●
PRESENT
STUDY
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 19 (a)

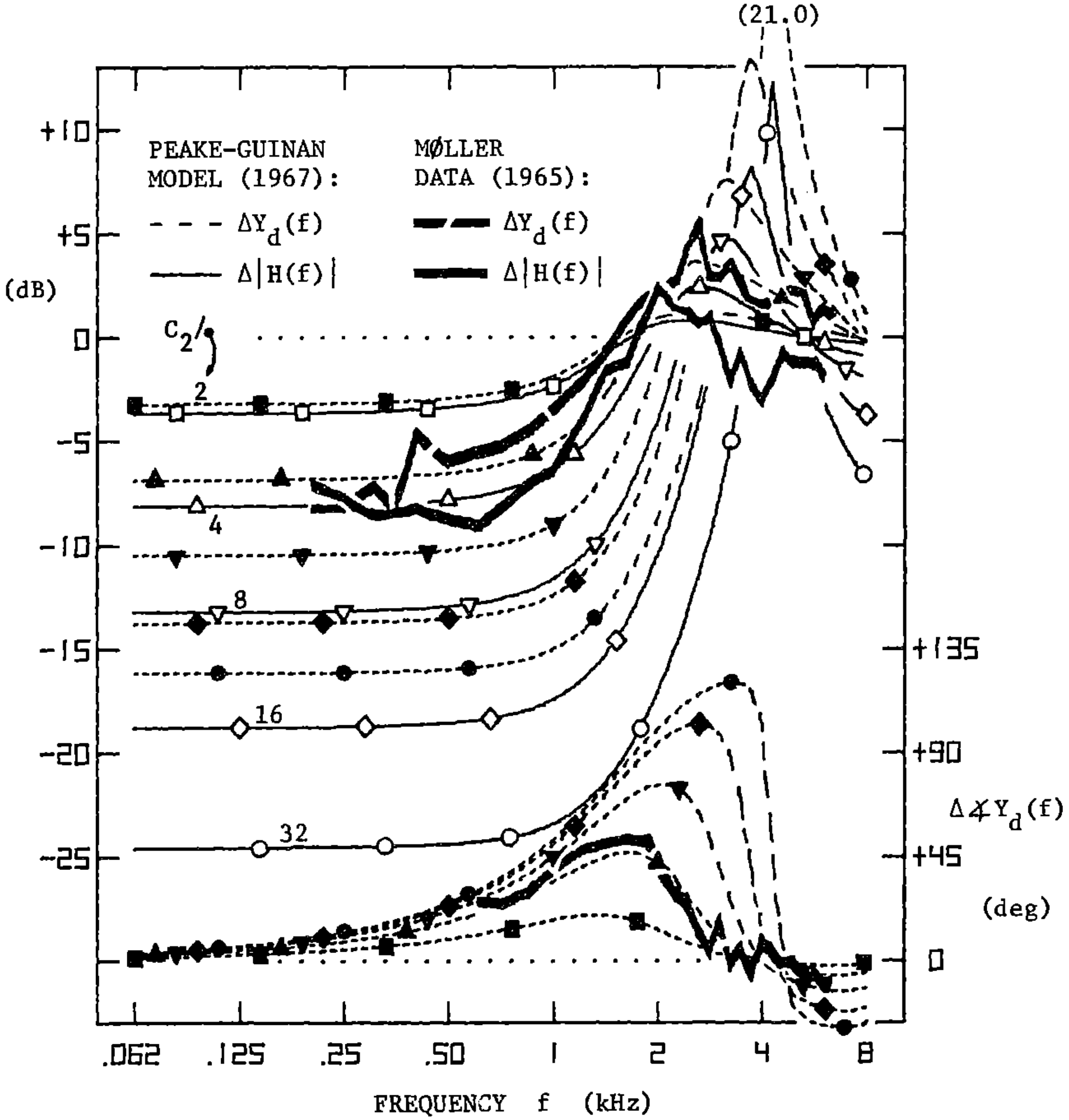

(21.0)
+10
+5
0
-5
-10
-15
-20
-25
(dB)
PEAKE-GUINAN MODEL (1967):
ΔY_d(f)
Δ|H(f)|
MØLLER DATA (1965):
ΔY_d(f)
Δ|H(f)|
C_2/
2
4
8
16
32
+135
+90
+45
0
Δ∡Y_d(f)
(deg)
.062
.125
.25
.50
1
2
4
8
FREQUENCY f (kHz)

FIGURE 19 (b)

## Appendix I. Alternate representations of our transmission-change results

Figure A-I. 1. Four-subject average transmission-change vs contralateral noise level with frequency, f, as a parameter. [Results replotted from Fig. 11.] Results for f = 400, 500, and 630 Hz averaged and indicated by $f \lesssim 0.63$ kHz.

Figure A-I. 2. Transmission-change data for each subject, (a) S1 to (d) S4, at each test frequency up to 1600 Hz. Straight-line fits shown as described in the text; fitted parameters are summarized in Table A-I. 1. Data were also taken for f = 2, 2.5, 3.15, and 4 kHz, but no effects on transmission occurred for any subject; i.e. for $L_N \lesssim 110$ dB SPL and $2 \lesssim f \lesssim 4$ kHz, all measurements of transmission-change were within ±1 dB of zero.

Table A-I. 1. (As described in table.)

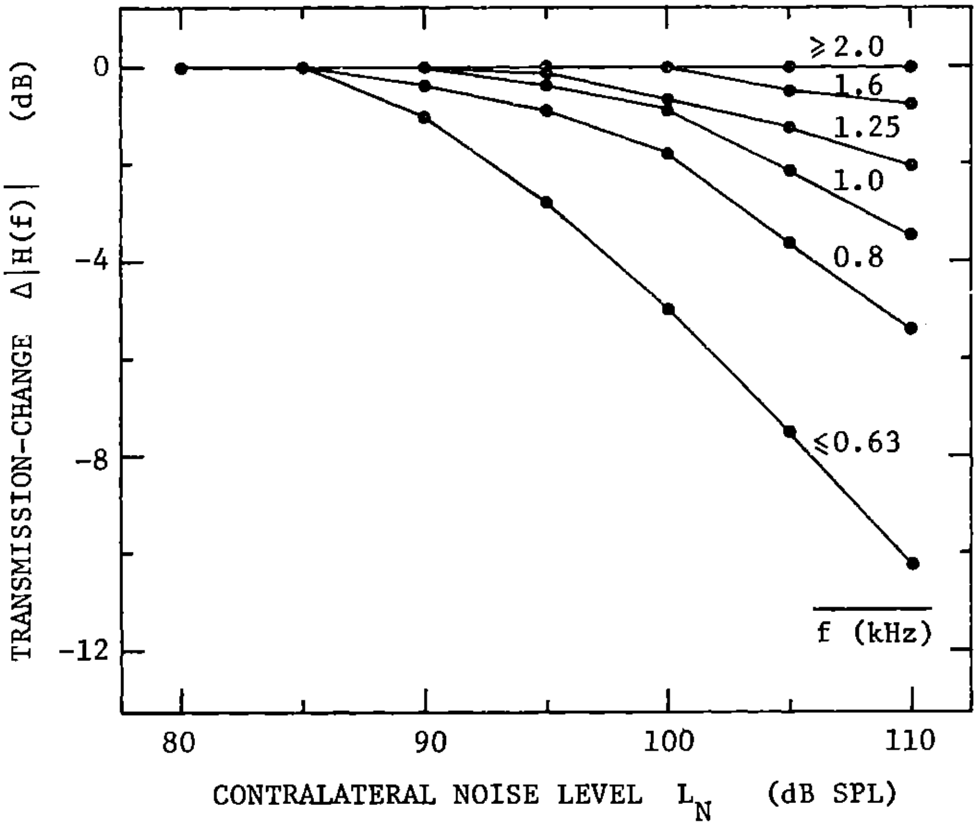
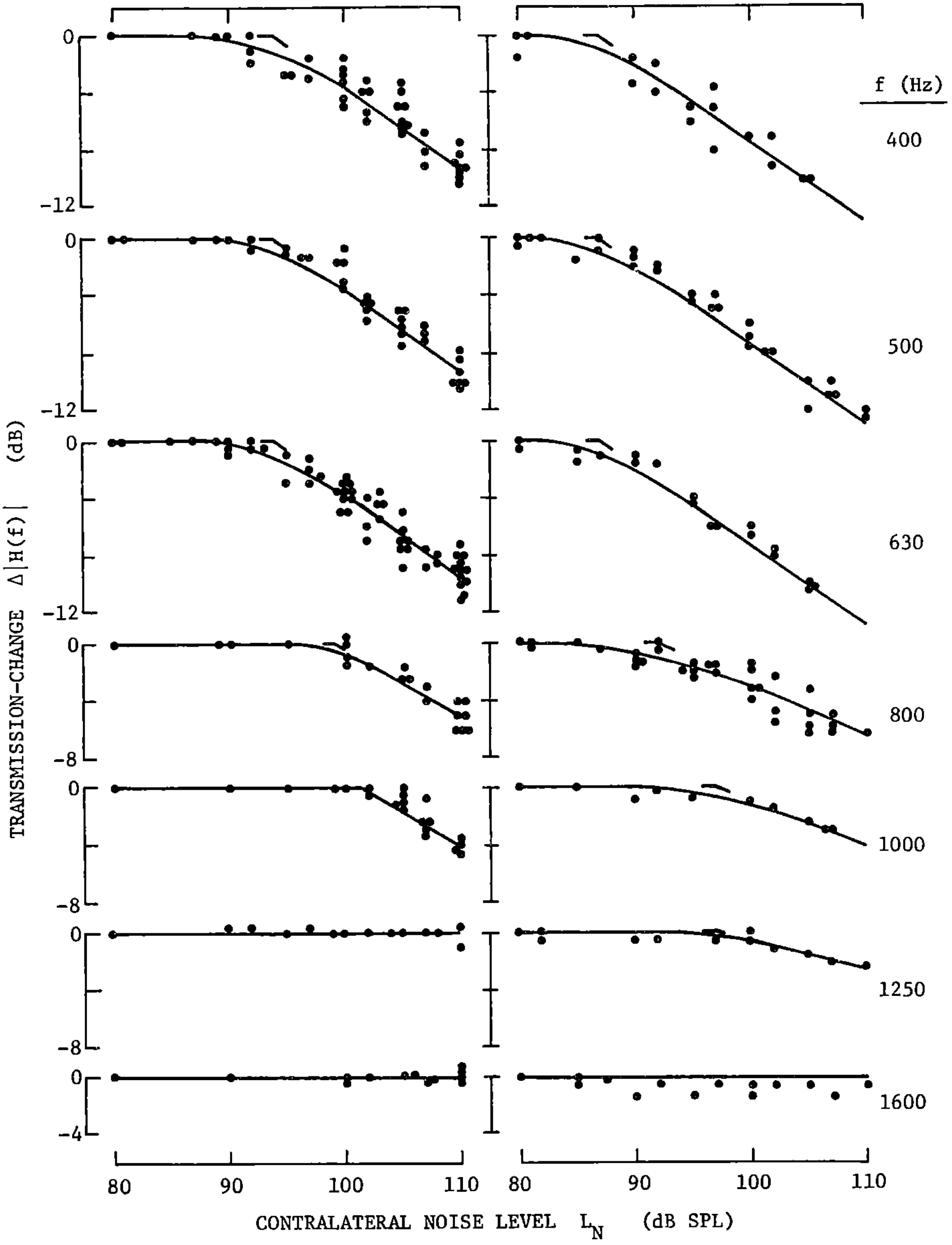
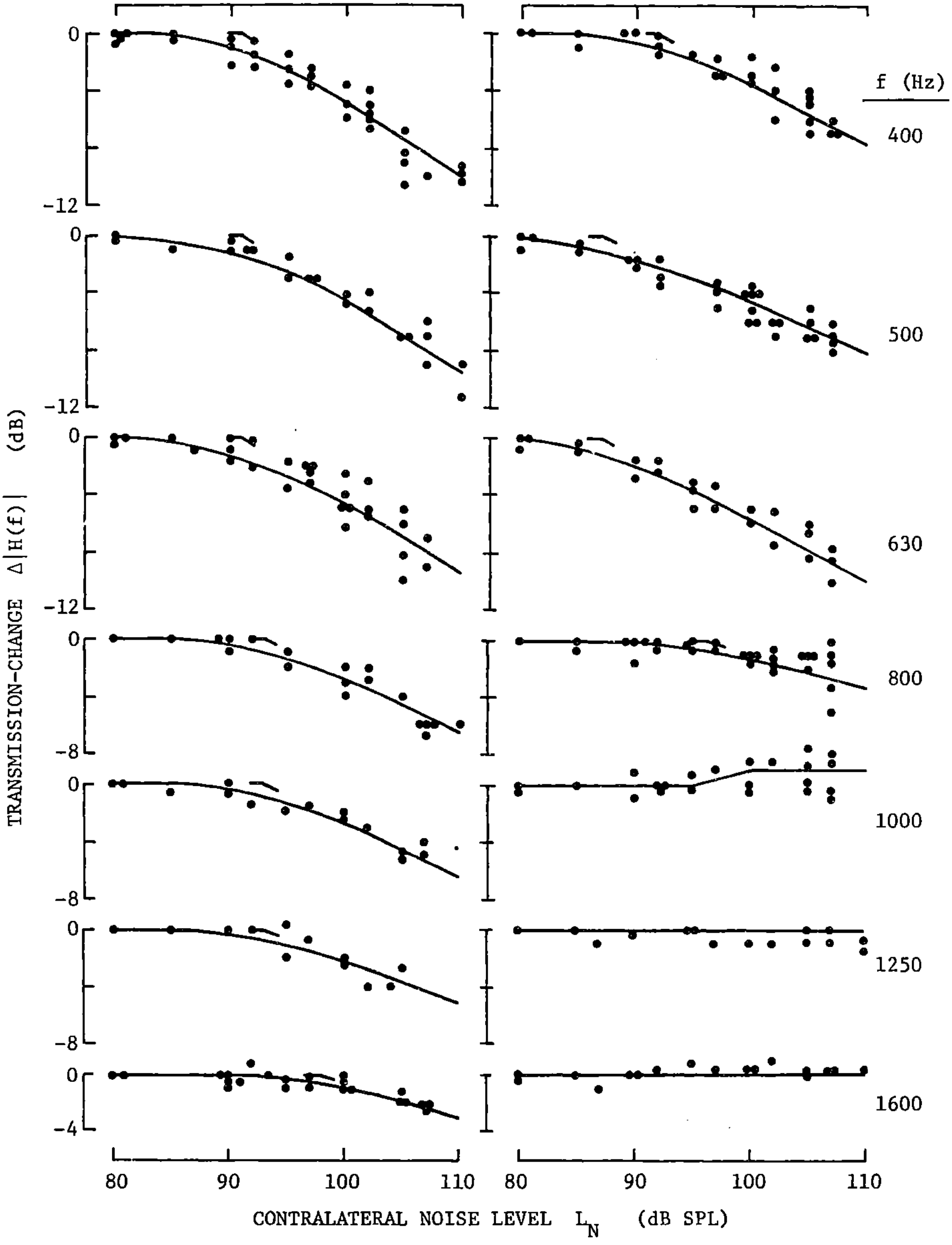

0
-4
-8
-12
80
90
100
110
TRANSMISSION-CHANGE Δ|H(f)| (dB)
CONTRALATERAL NOISE LEVEL $L_N$ (dB SPL)
⩾2.0
1.6
1.25
1.0
0.8
⩽0.63
f (kHz)

FIGURE A-I.1



(a) S1 (b) S2

FIGURE A-I.2 (a and b)



(c) S3

(d) S4

FIGURE A-I.2 (c and d)

(Perlman and Case, 1939; Fisch and Schulthess, 1963; Djupesland, 1965, 1967; Zakrisson et al, 1974).

Second, eardrum impedance-change studies have been done on patients with: (a) unilateral Bell's palsy which presumably renders the stapedius muscle paralyzed, (b) unilateral ossicular discontinuity (fracture of the long process of the incus) leaving the stapedius muscle operational but unable to influence eardrum impedance-change, (c) post-operative stapedectomy with the stapedius muscle having been sectioned, (d) post-operative exploratory tympanotomy with the stapedius tendon having been sectioned, and (e) unilateral footplate otosclerosis [(a) Borg, 1968; (a,c,d) Feldman, 1967; (a) Jepsen, 1955; (a,b,c) Klockoff and Anderson, 1960, and Klockoff, 1961; (a,e) Lidén et al, 1970; (c) Mills and Lilly, 1970]. These patients exhibited no impedance-change for the affected ear in response to intense contralateral acoustic stimulation.[1] The eliciting stimuli varied among the studies both in type, tones or noise, and maximum level, 110 to 125 dB SPL, but were sufficient in all cases to cause substantial or maximum impedance-change in normal ears. The tensor tympani was presumed functional in all these cases, and in the studies by Klockoff (1961) and Lidén et al (1970), the tensor tympani was demonstrated to be functional by observing impedance-change to non-acoustic elicitors. In contrast to this lack of impedance-change to contralateral acoustic stimulation when the stapedius is "absent" or nonfunctional, patients with a nonfunctional tensor tympani do show impedance-change (Metz, 1946; Lindström and Lidén, 1964; Feldman, 1967).

Table A-I. 1. Fitted parameters to the individual-subject AR data. Entries for each subject, S1 to S4, are from the straight-line fits to the raw $\Delta|H|$ data (see Fig. A-I. 2), and are the value of contralateral noise intensity $L_N$ (dB SPL) for the onset of AR-induced attenuation and the "regulation efficiency" (dB of transmission reduction per dB increase in $L_N$). The data for S4 at 1 kHz (*) were fit with $\Delta|H| = 0$ for $L_N \leq 95$ and $\Delta|H| = +1$ dB for $L_N \geq 100$ dB SPL. The dashes indicate $\Delta|H| = 0$ for all values of $L_N$.

| f (Hz) | S1 | S2 | S3 | S4 |
|---|---|---|---|---|
| 400 | 94 , 0.58 | 87 , 0.56 | 91 , 0.53 | 92 , 0.43 |
| 500 | 94 , 0.57 | 87 , 0.56 | 91 , 0.50 | 87 , 0.35 |
| 630 | 94 , 0.60 | 87 , 0.56 | 91 , 0.50 | 87 , 0.43 |
| 800 | 99 , 0.46 | 92 , 0.36 | 93 , 0.38 | 96 , 0.24 |
| 1000 | 101 , 0.45 | 97 , 0.33 | 93 , 0.38 | * |
| 1250 | - | 97 , 0.21 | 93 , 0.30 | - |
| 1600 | - | - | 98 , 0.25 | - |

## Appendix II. The acoustic reflex = the stapedius reflex?

Animal experiments involving sectioning of the stapedius and/or tensor tympani tendons indicate that contractions of the stapedius muscle are predominantly responsible for the transmission-changes accompanying acoustic-reflexes when both MEMs are intact. These experiments include direct measurements of transmission-change, which use the cochlear-microphonic-potential as a relative index of middle-ear output, on cat (Galambos and Rupert, 1959; Simmons, 19591 Irvine, 1976) and rabbit (Borg, 1972a), and indirect measurements in which transmission-change is inferred from contralateral eardrum impedance-chnage in rabbit (Borg, 1972c).

The human acoustic reflex also appears to be primarily a stapedius response. Tensor tympani involvement sometimes occurs, but usually at elicitor levels above those required to activate the stapedius. [Tensor tympani contraction does occur as part of a generalized "startle" or defensive reaction.] Two classes of evidence are relevant:

First, electromyographic studies (EMG) have demonstrated no tensor tympani EMG in response to (contralateral) tones or clicks at 110 dB SPL (Salomon and Starr, 1963) or wide-band noise at 134 dB SPL (Djupesland, 1965, 1967) that were not associated with simultaneous contractions of the muscles of the head and neck. In contrast, stapedius EMG activity in response to contralateral acoustic stimulation at these (and lower) levels has been recorded

Despite the above results, evidence for acoustically-elicited tensor-tympani contractions in a small fraction of normal subjects (13%) has been claimed by Lidén et al (1970) who employed a form of extratympanic manometry. [See also Terkildsen (1960) and Weiss et al (1963).] The basis for this technique is that MEM contractions cause eardrum movements. These (volumetric) movements can be inferred from pressure changes within a closed system sealed into the earcanal. Lidén et al adopted a criterion that "sudden spike-like negative deflection" (inward eardrum movement) which was "increasing with intensity" of the contralateral eliciting stimulus was indicative of tensor tympani contraction, whereas when the response was "either negative or positive, but the pattern broad and shallow" the interpretation was for stapedius contraction alone. Measurements were made on 127 ears of 78 normal subjects using various single-tone elicitors (of 1 sec duration). In all, 640 conditions (frequencies by ears) were tested. Tensor tympani responses, as defined by the above rule, were observed in 13% of the subjects, but only on 4% of the 640 conditions and the responses generally occurred at elicitor levels above those required to elicit stapedius responses. It was noted that no other facial musculature contractions or "intolerance" appeared to accompany these responses; hence, the responses were concluded to be acoustically elicited. Nevertheless, we suggest that the limited tensor tympani involvement concluded from these tests must still be regarded as conjectural because direct relationships between specific muscle actions and the recorded pressure changes have not been demonstrated.[2]

Furthermore, Lidén et al reported that no manometric or impedance-change responses were observable on patients (N = 39) with a nonfunctional stapedius muscle.

In summary, the above data suggest that the acoustic reflex is primarily a stapedius reflex in human, cat, and rabbit. [According to Perlman (1960), the guinea pig has no acoustic reflex.]

---

Footnotes for Appendix II

1. Borg (1968) observed small impedance-changes for some affected ears (Bell's palsy) at very high elicitor levels.

2. See also Mendelson (1966) for similar cautions on interpreting the manometric responses reported by Terkildsen (1960) and Weiss et al (1963).

# Appendix III. Measurements of middle-ear transmission-change in animals

The influence of MEM contractions on middle-ear transmission has been examined in numerous studies on several species. In most cases, the measurements have made use of the cochlear-microphonic-potential (CM) as a relative measure of middle-ear output. Transmission-changes are generally assumed to be equal to changes in the ratio $|CM/P_d|$ where $P_d$ is the sound pressure at the eardrum.

Two aspects of these data are considered below: (a) the magnitude of the transmission-change and its frequency dependence for a given state of muscle contraction, and (b) the growth of transmission-change as a function of elicitor level (for acoustic reflexes). Because the acoustic reflex is primarily a stapedius reflex (see App. II), we include in (a) data on the effects of the stapedius muscle acting alone.

## (a) Magnitude and frequency dependence of stapedius effect

Measurements of the effects of stapedius contractions on transmission-change are shown in Fig. A-III. 1. In addition to the results from cat, rabbit, and guinea pig, our results for the human acoustic-reflex elicited by 110 dB SPL contralateral noise are shown (from Fig. 11). Some experimental parameters for the studies are given in Table A-III. 1. The results shown can be placed into two groups. One group, including all but the lowest two curves, exhibits low-frequency attenuations of 5 to 15 dB and little or no effect for high-frequencies (> 2 kHz). The other group shows larger low-frequency attenuations (20 to 30 dB) which decrease with increasing frequency, but attenuations of 5 to 10 dB remain up to 10 kHz. It should be noted that all of the results in Fig. A-III. 1 derive from experiments in which moderate-intensity stimuli were used (i.e., $P_d \lesssim 80$ dB SPL).

Although measurement of $|CM/P_d|$ for a constant level of stapedius contraction is common to all of these animal studies, differences exist in the physiological state of the animal, the method by which the stapedius action was caused, and the level of the contraction. Examination of this last parameter, the level or strength of stapedial contraction, helps to interpret the range of results displayed in Fig. A-III. 1.

On anesthetized cats, Wever and Bray (1942) cut the stapedius muscle and applied tensions to the portion of the stapedius tendon still attached to the stapes. Care was taken to align the tensioning string so as "to give a normal direction of pull". The large effects shown for an applied tension of 11 g were thought to be in excess of normal physiological limits as, until recently, the cat stapedius muscle was thought to be capable of developing maximal tensions of only 1 g (Wever and Lawrence, 1954), 1.2 g (Wever et al, 1955), and 3.5 g (Wersäll, 1958). However, Teig (1972) has shown maximal stapedial tensions of 13.9 g for the cat (and 15.4 g for the rabbit, in contrast to an earlier estimate 8.6 g by Wersäll, 1958). In Teig's experiments, the facial nerve was electrically stimulated and the tension developed in the contracted stapedius muscle was measured with an electromechanical transducer connected to the lenticular process (i.e., the neck) of the stapes with the stapes crura and the incus removed. The larger tensions recorded (re older studies) were attributed by Teig partly to differences in muscle stimulation parameters, but primarily to the finding that developed tension was critically sensitive to the initial resting length of the stapedius muscle. Teig's tensions were given for an "optimal" resting length for which maximal tensions were obtained for single-shock twitch contractions. Using similar procedures, Teig (1973) recorded changes in $|CM/P_d|$ for seven levels of tetanic electric stimulation applied in the facial nucleus of the cat. After these measurements, the middle-ear cavities were opened widely and stapedius-tension measurements were made, at "optimal" muscle length, for the same seven electric stimulation levels. Fig. A-III. 1 shows Teig's data for a small tension of 1.04 g (this being level two, level one was 0.9 g) and for the largest tension of 5.17 g. The 5.17 g tension produced effects similar to the Wever and Bray (1942) data of 11 g of tension, except for the somewhat larger low-frequency attenuation in the Wever and Bray data. Some of this difference may derive from Teig's having used a closed-bulla preparation, whereas Wever and Bray used an open-bulla preparation, and also, inter-animal variability could be important since the results are reported for only one animal. For tensions between the two values shown, Teig observed that transmission changed monotonically with tension. At low frequencies, it was noted that the rate of transmission-reduction decreased with increasing tension (i.e., incremental tensions caused greatest attenuation starting at zero muscle tension, see also Wever and Bray, 1942).

For the remaining two studies on the cat, Wever and Vernon (1955) and Møller (1965), contractions of the stapedius were elicited by contralateral acoustic stimulation (i.e., an acoustic reflex). Tensor tympani involvement was eliminated by severing the tensor tympani tendon (Wever and Vernon) or by the application of a local anesthetic to the muscle (Møller). Wever and Vernon used awake-decerebrate cats, and a reflex-elicitor (1 kHz) only 2 dB above the reflex threshold. Møller used an intense elicitor, 125 dB SPL (800 Hz), on a cat "lightly" anesthetized with phenoperidine. In both cases the results are similar to Teig's 1.04 g results; we conclude both procedures resulted in small tensions being produced.

In studies on rabbit (Møller, 1965) and guinea pig (Nuttall, 1974), electric stimulation to the stapedius muscle was used. On the rabbit, Møller caused twitch contractions with single shocks applied "on, or near, the tendon of the stapedius muscle". As the tetanic-to-twitch tension ratio (for maximal contractions) is 2.2 for the rabbit stapedius (Teig, 1972; Wersäll, 1958), the capability of the stapedius muscle in reducing low-frequency transmission for tetanic stimulation is likely to exceed the -13 dB shown. For the guinea pig, Nuttall applied stimulation to the stapedius muscle body. To normalize across animals, Nuttall adjusted current levels to produce the same transmission-change at 300 Hz. Note that although the results shown are reported to be "near-maximal (stapedial) contraction", only 10 dB reductions occur at low frequencies, in contrast to the 20 to 30 dB for maximal contractions in the cat.

### (b) Dependence on elicitor level

Using paradigms functionally equivalent to that used in our psychophysical studies, the growth of induced attenuation for the acoustic reflex has been measured by Borg (1972a) and Price (1966) on rabbits "lightly" anesthetized with sodium pentobarbital, and by Wever and Vernon (1955) on decerebrate cats. The transmission-change, $\Delta|CM/P_d|$, imposed on a moderate-intensity tone applied to one ear was measured as a function of the intensity of a reflex-eliciting tone applied to the contralateral ear. For low-frequency tones in the measurement ear, the reflex-induced attenuation in dB was observed to grow almost linearly with the intensity in dB of the contralateral eliciting-stimulus over a range of 20 to 30 dB for the eliciting stimulus. Borg's data (1972a, Figs. 4 and 5) are fit by a growth-rate of 0.43 dB/dB (i.e., ~0.5 dB/dB for the dashed line in the left panel of Fig. 4 and ~0.35 dB/dB for the dashed line in the right panel of Fig. 5). [For these experiments a 2-kHz elicitor was used and $|CM/P_d|$ was measured at 800 Hz.] In contrast, Price (1966, Fig. 2) and Wever and Vernon (1955, Fig. 3) observed a growth-rate of 1 dB/dB up to a maximum, saturating value of attenuation of 20 dB. This growth-rate implies that "perfect regulation" occurs, whereby middle-ear output is maintained constant while eardrum sound-pressure increases over a 20 dB range. Perfect regulation over that large a range is, however, doubtful; we suggest, these large growth-rates occurred because of an asymmetry in the state of the middle-ear cavities on the elicitor- and measurement-ears, coupled with the particular stimulus frequencies used.[1]

Simmons (1963) on the cat used one ear for both reflex-elicitation and transmission-change measurement. An electrode was (chronically) implanted on the round-window, the bulla was resealed, and the animal was then allowed to recover from the surgery. Measurements were then made for $|CM|$ at 1 kHz as a function of $|P_d|$. Two such functions were obtained, one with the cat awake and one with the cat anesthetized. It was assumed that "(at 1 kHz) virtually any amplitude difference between CM measurements from awake cats and those from anesthetized cats is due to activity of the middle ear muscles" in the awake cat (Simmons, 1963, pg. 530, based on results in Simmons, 1959). Thus, subtraction of $|P_d|$ (in dB SPL) in the anesthetized cat from $|P_d|$ (in dB SPL) in the awake cat which both produced the same $|CM|$ yielded estimates of reflex-induced attenuation as a function of $|P_d|$ in the normal, awake cat. The data[2] (Simmons, 1963, Fig. 2, 16 cats) indicate large interanimal differences, but the median attenuation increase almost linearly from 5 to 35 dB as $|P_d|$ increases from 45 to 115 dB SPL, for a "regulation efficiency" of 0.43 dB/dB.[3]

With an indirect procedure making use of eardrum impedance-change as a relative index of contralateral middle-ear output, Borg (1972c) studied reflex-regulation in the nonanesthetized rabbit. [The procedure has also been applied to studies of the human acoustic-stapedius relfex (Borg, 1968; Borg and Zakrisson, 1974) and is described elsewhere, see Sec. IV. C. (2.a).] For the "total" reflex (i.e. with both MEMs functional) the regulation measured was about 0.7 dB/dB. With the stapedius denervated, a regulation of 0.2 to 0.3 dB/dB was measured for the tensor tympani. However, with the stapedius intact, no effect on regulation was measureable when the tensor tympani was sectioned. Thus, it was concluded that the regulation in the normal middle-ear is almost entirely attributable to the stapedius muscle.

The above growth-rates of attenuation can be identified in terms of the regulation measures discussed in Sec. IV. C. (2.b). The values of 0.43 dB/dB for the cat from Simmons (1963) and 0.7 dB/dB for the rabbit from Borg (1972c) are estimates of $\eta_i$ as these measures indicate transmission-changes in dB per dB increase in sound pressure to the regulating ear. The value of 0.43 dB/dB for the rabbit from Borg (1972a) is not explicitly $\eta_i$ or $\eta_o$ because (as in our tests) transmission-change was measured at a low frequency (800 Hz) vs the level of an eliciting stimulus of higher frequency (2 kHz). However, (a) insofar as the 2-kHz elicitor is unaffected by the ipsilateral reflex action it elicits, at least up to 15 to 20 dB re reflex threshold (Borg, 1972c), and (b) insofar as the "light" anesthesia with sodium pentobarbital used by Borg (1972a) had no effect on the growth of MEM response, except for a possible shift in the reflex threshold [e.g. compare the slopes of impedance-change vs elicitor level in Borg and Møller (1975) to those in Borg (1972b, 1972c)], we suggest that Borg's (1972a) experiment in rabbit is functionally equivalent to ours.

## Footnotes for Appendix III

1. For example, Wever and Vernon (1955) used a 1-kHz elicitor and measured contralateral $|CM/P_d|$ at 300 Hz. Let us suppose for the cat that (a) a given contraction reduces transmission uniformly at 300 Hz and 1 kHz, and (b) the reflex exhibits perfect bilateral symmetry with equal signals being sent to the MEMs of both ears. Then, for the two bullas being either both open or both closed, the reflex-attenuation and its growth-rate would be equal in both ears. However, Wever and Vernon (and also Price, 1966, on the rabbit) had the bulla closed on the elicitor ear, but open on the measurement ear. Because the closed-bulla provides added stiffness (at low frequencies) to the middle-ear system of the cat (c.f. Møller, 1965; Guinan and Peake, 1967; and for rabbit, Møller, 1965; Borg, 1972a), the induced-attenuation and its growth-rate will be greater in the open-bulla ear than in the closed-bulla ear.

2. The 1 kHz tones were 60 sec in duration and reflex-attenuation was calculated ~2 sec after the tone onset and again near the end of the 60-sec stimulus period. The later measurements show less attenuation than the early ones, indicating adaptation of the reflex response. The discussion in the text refers to the early measurements.

3. The value of ~45 dB SPL at which the reflex-attenuation begins in Simmon's cat data is significantly lower than that measured in other species. Our results show little attenuation occurs for elicitors below 90 dB SPL, and Borg (1972c) on the nonanesthetized rabbit shows little attenuation below 80 dB SPL. However, in a review of species differences in cat, rabbit, and man for the excitability of middle-ear reflexes, Borg (1972b) has noted that for the cat there is (a) wide variability in reported reflex thresholds, (b) poor reproducibility of measurements over days or months, and (c) additional instabilities owing to habituation and rapid adaptation. Borg (1972b, pg. 386) concluded that given all the available data "there is little reason to postulate that the average excitability values differ significantly between the cat and the rabbit.

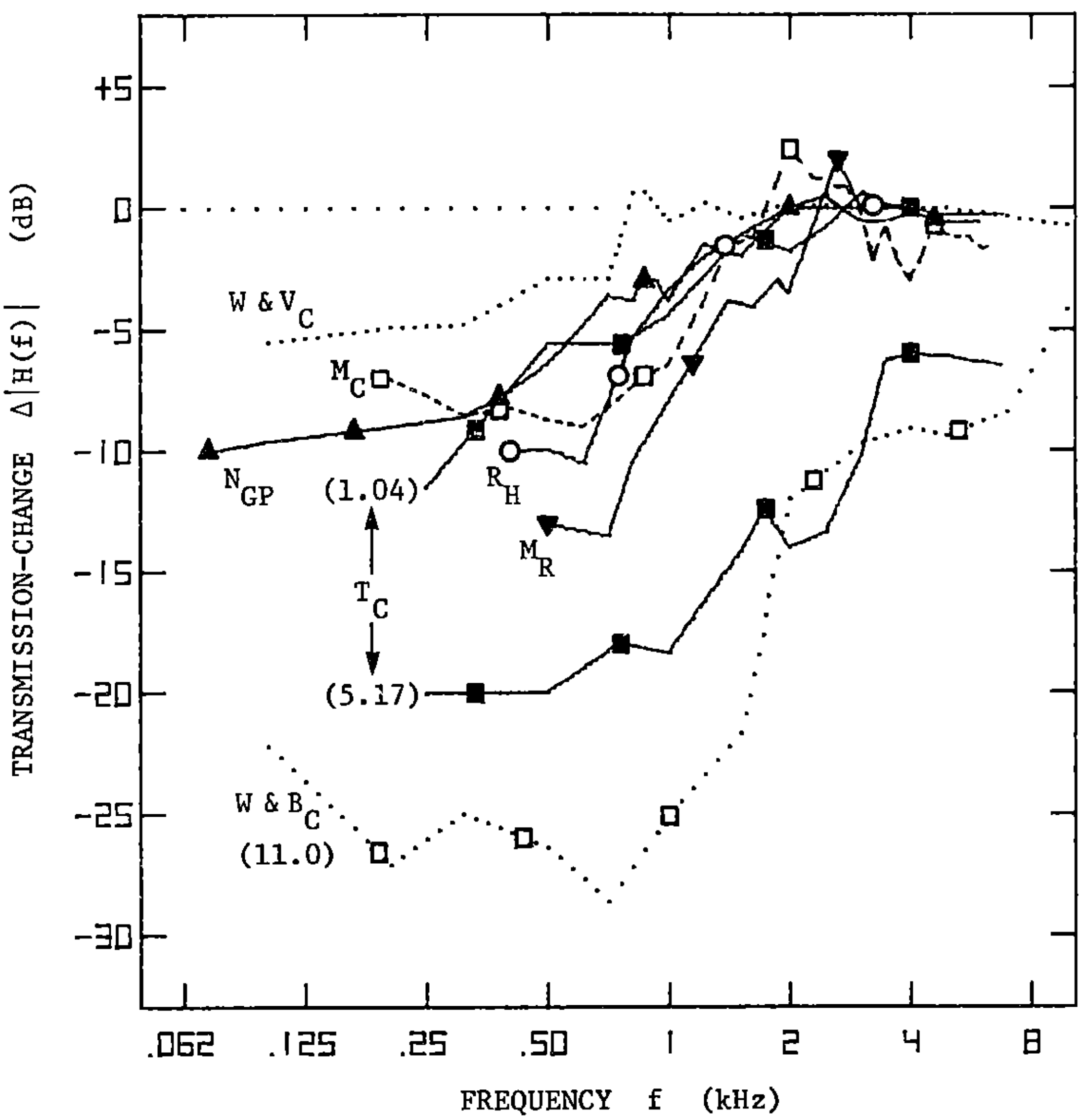


Figure A-III. 1. Transmission-changes for constant levels of stapedius contraction reported in certain animal studies, and our results for the human acoustic reflex with $L_N$ = 110 dB SPL. Labels alongside the low-frequency portion of each function are the initials of the authors (see Table A-III. 1), subscripts denote the species tested (C = cat, GP = guinea pig, H = human, and R = rabbit), and the numbers in parentheses for $T_C$ and W & $B_C$ are tensions in grams.

Table A-III. 1. Summary of the experimental parameters for the results in Fig. A-III. 1

| Study | Specie (Number) | Middle-ear Cavities | Anesthesia | Type of Stimulation |
|---|---|---|---|---|
| Møller (1965, Fig. 14) | Cat (1) | Bulla and septum open | Phenoperidine ("light") | 800 Hz contralateral tone at 125 dB SPL. |
| Møller (1965, Fig. 15) | Rabbit (1) | Bulla open | Urethane or sodium pentobarbital | Single shocks to stapedius causing twitch contractions |
| Nuttall (1974, Fig. 4) | Guinea pig (15) | Bulla open | Dial | Tetanic electric stimulation of stapedius causing "near-maximal" contraction |
| Teig (1973, Fig. 3a) | Cat (1) | Closed | Sodium pentobarbital | Tetanic electric stimulation applied in facial nucleus |
| Wever and Bray (1942, Fig. 4) | Cat (1) | Bulla and septum open | Dial and curare | Tension applied to stapedius tendon |
| Wever and Vernon (1955, Fig. 2) | Cat (1) | Bulla and septum open | None (decerebrate) | 1 kHz contralateral tone at +2 dB re reflex-threshold |
| Rabinowitz (Fig. 11) | Human (4) | Intact | None | 2 to 4 kHz contralateral noise at 110 dB SPL |

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