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A cancellation technique for the amplitude and phase calibration of hearing aids and nonconventional transducers*

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BACKGROUND

A difficult problem in specifying the characteristics of electroacoustic transducers is the fact that the sound pressure level measured in a standard coupler or artificial ear is not the same as that at the eardrum. A related problem is encountered in specifying the overall gain of a hearing aid. This latter problem is especially difficult because of the complexity of the acoustic coupling between the hearing aid receiver and the eardrum. An additional complicating factor is the acoustic effect of mounting a microphone on the body (usually on the ear).

The above problems are compounded by the difficulty of measuring acoustic power (or energy) directly. Whereas sound pressure is relatively easy to measure, it is much more difficult to obtain measurements of volume velocity. In order to determine the power of a sound, however, it is necessary to know both the pressure and the volume velocity, as well as the relative phase of these two variables. Alternatively, the power of a signal can be derived from measurements of sound pressure and acoustic impedance. The latter variable is, in general, a complex quantity with both real and imaginary components and it is also not easily measured.

In view of the above difficulties, it is common practice to specify the relative strength of an acoustic signal in terms of its pressure level (e.g., sound pressure level, sensation level, hearing level). This approach is quite satisfactory provided the system or conditions being compared have the same acoustic impedance, as is often the case in practice. Care must nevertheless be exercised in interpreting pressure level measurements. For example, auditory thresholds are typically measured in terms of just-detectable pressure levels; consequently, differences in measured thresholds between normal-hearing and hearing-impaired persons with middle ear pathologies reflect not only differences in auditory sensitivity but also the differences in acoustic impedance between the normal and impaired ear. Similarly, the variation in auditory threshold as a function of frequency is a result of variations in both auditory sensitivity and acoustic impedance as a function of frequency.

A matter of some concern in specifying the effective gain of a hearing aid is that the two conditions of interest (aided vs unaided) involve different forms of acoustic coupling and in general different acoustic impedances. For example, the transfer function specifying the change in sound pressure level from sound field to eardrum does not cover all of the factors affecting the detectability of the amplified sound. A more reliable method is to specify the change in pressure level at a common point (e.g., at the eardrum) resulting from the use of the hearing aid.

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THE FUNCTIONAL-GAIN TECHNIQUE

A very powerful technique for circumventing the above problems is that of measuring the *functional gain* of a hearing aid (11). According to this technique the threshold of audibility for a frequency-specific signal (tone or narrow band of noise) is obtained for both the aided and unaided conditions. The ratio of these two thresholds (in dB) is defined as the functional gain of the hearing aid at the test frequency. Since detectability is measured directly for the two conditions of interest, all of the factors affecting the gain of the hearing aid (e.g., hearing aid transfer function, acoustic coupling effects, individual differences in acoustic impedance) are taken into account.

The functional gain technique is limited in several ways. The most obvious is that the key measurement, the threshold of audibility, is a psychophysical rather than a physical measurement and is thus subject to all of the limitations of psychophysical testing; e.g., test-retest variability is typically much larger than that obtained with physical measurement and the time required per measurement may be orders of magnitude greater (minutes as compared to seconds). Despite these limitations, the technique is nevertheless sufficiently practical for laboratory or clinical use and is worth the additional effort if accuracy rather than precision of measurement is the primary concern.

Measurements of functional gain are usually obtained in the sound field, and in order to avoid the problem of standing waves in the test room it is common practice to use narrow bands of noise rather than tones as the test stimuli. One-third-octave bands of noise are often used for this purpose, and it is important to bear in mind that the skirts of conventional one-third-octave band filters are fairly broad for attenuations in excess of 30 dB.

The skirt of the noise band presents a problem when measuring functional gain for a subject with a steeply sloping hearing loss, since energy in the skirt of the noise band may be detected in an adjacent frequency region where residual hearing happens to be relatively good, rather than the main body of the noise being detected in the frequency region of interest. If a very narrow band of noise is used, the random variations in the amplitude of the noise band will be large, resulting in a high test-retest variability. This problem is not unique to the measurement of functional gain, but applies to audiometric sound-field measurement in general. The problem of stimulus variability can be avoided by using a non-random narrowband stimulus such as a frequency-modulated tone with strictly controlled bandwidth and rate of modulation.

Another practical problem is that the internal noise of the hearing aid may exceed the subject's threshold of audibility, in which case the measured threshold for the amplified condition will be raised. Since only the aided threshold is affected, the measurement of functional gain under these conditions will underestimate the true gain of the hearing aid.

A third practical problem is that the transmission path from sound field to hearing aid microphone is critically dependent on head position and on the exact location of the microphone on the ear (7). A small change in the location of the microphone, or a movement of the head, can change the functional gain of a hearing aid significantly. As before, this problem is not so much a limitation of the functional-gain technique *per se*, but is an inherent problem for all methods of measuring hearing aid gain. Specifically, the overall gain of a hearing aid is a function of relative head position and how the hearing aid is mounted on the ear (or elsewhere on the body).

THE REFERENCE HEARING AID TECHNIQUE

A technique that is similar in concept to the functional-gain method, but which avoids most of the above problems, is the *reference hearing aid* technique. This technique was designed to facilitate measurement of hearing aid characteristics in studies involving adjustable or programmable hearing aids (8,9). In this procedure, the overall gain of the hearing aid is subdivided into three components:

1) the transfer function from sound field to electrical output of the microphone;

2) the transfer function from electrical input to the hearing aid to the electrical output of the hearing aid (i.e., the electrical signal delivered to the hearing aid receiver);

3) the function relating electrical input to the hearing aid receiver to the subject's behavioral threshold.

The first of these transfer functions is the most

variable. It is usually measured for a single reference condition, e.g., the listener facing the sound source at 0 degrees of azimuth at a distance of 2 meters. Additional measurements are also sometimes obtained showing the range of variation of this transfer function for changes in head position typical of those occuring in face-to-face communication. The transfer function for the reference condition is used in deriving the overall gain of the hearing aid; variations in the transfer function about this reference condition provide information on the changes in hearing aid gain to be expected from changes in head position relative to the sound source.

The second transfer function is the easiest to measure and can be measured most precisely. Electrical measurements only are involved and a signal level well above the internal noise of the hearing aid is normally used. As before, measurements are obtained for a reference condition. For convenience, a flat frequency response at a typical average gain (e.g., 30 dB) is used for the reference condition.

The third function, relating electrical input at the hearing aid receiver to the subject's threshold of audibility, covers a number of complex effects. These include electrical to acoustical transduction in the hearing aid receiver, the transmission characteristics of the sound path from hearing aid receiver through the acoustic tubing, earmold, and earcanal to the eardrum, the transduction from acoustic to mechanical vibration at the eardrum, the sound-transmission characteristics of the middle ear and cochlea and subsequent processing of the signal at higher levels in the auditory system leading finally to a behavioral response. Although complex, the function relating electrical input at the hearing aid receiver to behavioral threshold is reasonably stable, provided an efficient and precise psychophysical test procedure is used; e.g., an adaptive up-down technique using a forced-choice paradigm.

Once the various transfer functions for the reference hearing aid have been obtained, it is a relatively simple matter to determine the effect of a change in hearing aid parameters on the threshold of audibility. This aspect of the technique is particularly useful when adjusting the parameters of a programmable hearing aid, since all subsequent measurements can be obtained electrically (including automated self-calibration using a computer-based system) and related directly to the signal levels obtained for the reference hearing aid condition. It is important, when using this technique, that the acoustic coupling in the hearing aid that is finally fitted to the subject be the same as that used in the reference hearing aid; i.e., the same receiver, earmold, and acoustic tubing should be used.

The reference hearing aid technique is not restricted to threshold measurements only. It is also used to determine the relationship between the acoustic or electrical input to the reference hearing aid and the subject's loudness discomfort level, as well as other loudness levels (9). The threshold data obtained with the reference hearing aid technique can be used to determine the changes in functional-gain resulting from changes in hearing aid parameters. The loudness data can be used to determine appropriate output-limiting levels and compression-amplification characteristics for the subject.

An important advantage of the reference hearing aid technique is that, by separating the overall gain into the three transfer functions, it is possible to obtain threshold data that are not contaminated by the internal noise of the hearing aid. It is also possible to obtain sound-field data separately so that transfer functions 2 and 3 can be obtained with tonal stimuli. The use of tonal stimuli not only increases precision but also allows the measurement of the phase response of the hearing aid.

MEASUREMENT OF PHASE-FREQUENCY RESPONSE

Phase Characteristics

The phase-frequency response of hearing aids has received very little attention. This is presumably because the phase characteristics of audio signals are often assumed to be of little practical consequence. Whereas the ear is relatively insensitive to phase in monaural listening, it is remarkably sensitive to phase in binaural listening. A normal-hearing listener, for example, can detect interaural phase differences as small as 2.5 degrees, depending on the frequency and sound pressure level of the signal (14). For a 1000 Hz tone at 65 dB SPL the just-noticeable-difference (JND) in phase is about 3.5 degrees—this corresponds to an interaural time difference of 10 microseconds, an extremely small difference by any measure.

Since the vast majority of hearing aids are de-

signed for monaural use, it is not surprising that the phase characteristics of hearing aids have received little attention. Even the so-called binaural hearing aid is typically no more than a pair of monaural hearing aids that have been fitted individually to each ear. The exquisite sensitivity of the binaural system to interaural phase differences is simply not being taken into account in either the design or fitting of binaural aids.

There are additional reasons for being concerned with the phase characteristics of hearing aids. Acoustic feedback can be reduced by appropriate manipulation of the phase characteristic of the hearing aid (2). It is also possible to reduce the peak factor (peak/rms) of amplified speech signals by manipulation of the phase characteristic. Reducing the peak factor can be used to reduce the amount of distortion produced by peak clipping, a technique often used to protect hearing aid users from occasional intense sounds.

Whereas it is relatively difficult to control phase precisely in conventional analog circuits, it is no more difficult to control the phase-frequency response of a digital filter than to control its amplitude-frequency response. It is likely that more attention will be paid to phase-frequency characteristics in the hearing aids of the future using digital or quasi-digital techniques. It is thus important that methods be developed for measuring and calibrating the phase of hearing aids.

Cancellation Technique

A very sensitive method of measuring relative phase is by means of cancellation (5). In this technique, the test tone delivered by the hearing aid is cancelled by a reference tone at the same frequency reaching the cochlea by an alternative route. The alternative route could be by bone conduction or by means of the electrophonic effect (12). In order to achieve cancellation it is necessary to adjust the amplitude and phase of either the test or reference tone is such a way that the signals reach the chochlea with equal amplitude, but opposite in phase.

The following example shows how this technique was used in checking the phase characteristic of a digital hearing aid. The hearing aid consisted of a conventional hearing aid microphone followed by a 32-tap finite-impulse-response (FIR) digital filter, power amplifier, limiting network, and output transducer. The hearing aid was first programmed to

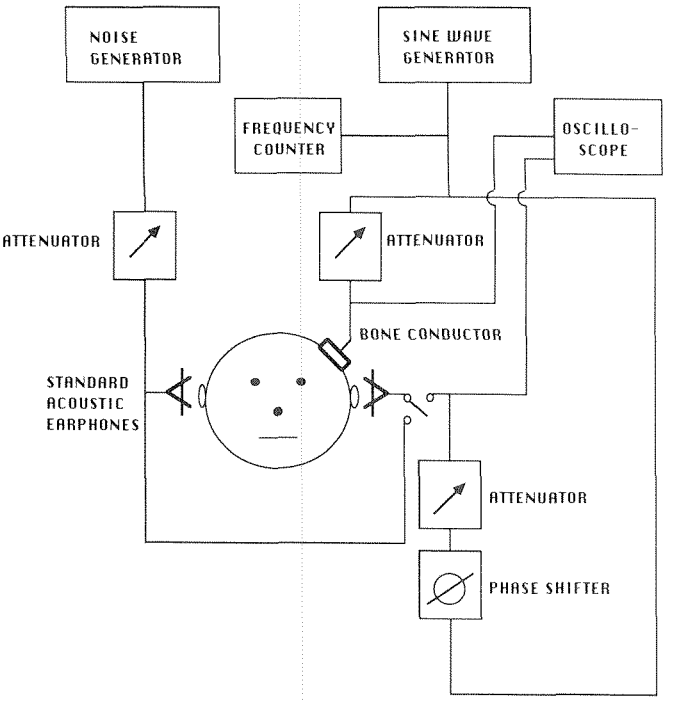

Figure 1.
Block diagram of experimental equipment.

have a flat electrical response. This was the reference hearing aid condition.

It was of interest to determine the phase frequency response at the ear for a nominal 90-degree phase shift in the digital hearing aid. The transfer function from sound field to microphone was not of interest in this example and the test signal was applied after the microphone at the electrical input to the hearing aid. The reference signal was delivered by means of bone conduction.

A block diagram of the apparatus is shown in **Figure 1**. A low distortion sine-wave generator was used as the signal source. An oscilloscope and frequency counter were used to monitor the output of the sine-wave generator. The signal was then split into two parts, a *test* tone and a *reference* tone. The latter signal was amplified, fed to a phase shifter and then to a precision attenuator before being applied to the reference transducer. The reference transducer in this example was a Radio Ear B-70A bone-conduction vibrator mounted on the mastoid. The test signal was led directly to the electrical input of the digital hearing aid, the output of which was delivered to the hearing aid receiver.

The hearing aid was adjusted to deliver the test tone at a comfortable listening level. The test tone

was then switched off and the bone-conducted reference tone switched on and adjusted in level to be slightly louder than the test tone. Since the reference tone was transmitted by bone conduction it was received by both ears, resulting in a binaural sound image located between the two ears. Masking noise was added to the non-test ear. As the level of the masking noise was increased, the perceived position of the reference tone moved toward the test ear. The masker level was increased until the reference tone reached the test ear and any further increase in the masker level produced no further change in the perceived position of the reference tone.

Both the test and reference tones were switched on and the amplitude and phase of the reference tone were then adjusted until the two signals cancelled each other. An iterative method of adjustment was used. The attenuator was first adjusted so that the reference tone, when heard alone, was roughly as loud as the test tone heard alone. The phase shifter was then adjusted until the combined tone reached a minimum in loudness. The amplitude of the reference tone was then adjusted to obtain a lower loudness minimum. The phase shifter was then adjusted to reach an even lower loudness minimum, and so on, until the combined tone was no longer audible. At this stage, the test and reference tones had effectively cancelled each other. The adjustment procedure was found to be both rapid and precise. Typically, only three or four adjustments were needed in order to achieve cancellation.

At cancellation, the rms voltage and relative phase of the reference tone were recorded. The measurements thus obtained provided baseline data for the reference condition. The digital hearing aid was then re-programmed to have the same amplitude-frequency response but with a 90-degree phase shift at all frequencies. The cancellation procedure was then repeated. The difference between the two sets of measurements shows the change in functional gain, for both intensity and phase, resulting from the re-programming of the hearing aid. The experiment was then repeated in order to obtain an estimate of test-retest variability.

The data are summarized in **Table 1**. The standard error was found to be 1.4 dB for the intensity measurements and 8.5 degrees for the phase measurements. Although the digital hearing aid was programmed for a nominal change of 0 dB in intensity and 90 degrees in phase for all frequencies, the data show systematic changes in both intensity and phase as a function of frequency. These differences were due to the limitations of the 32-tap FIR digital filter. Electrical measurements of the gain of the digital hearing aid show changes in intensity and phase with frequency similar to those obtained for the measured functional gain.

Figure 2 shows the measured changes in intensity and phase for the digital hearing aid. The solid curves show the electrical measurements, the plotted points show the changes in gain as measured using cancellation. Both sets of data show similar changes in intensity and phase with frequency. The standard deviation between the functional gain measurements and the much more precise electrical measurements was 2.4 dB for intensity and 11.6 degrees for phase. These standard deviations are consistent with the estimated test-retest variability of the functional gain technique. The corresponding standard errors of the electrical meaurements were 0.1 dB for intensity and 2 degrees for phase, respectively.

Table 1.
Measured change in functional gain

		250	500	1000	2000	4000
Intensity change (dB)	Replication 1	−2	0	−1	−2	5
	Replication 2	−2	−5	−1	1	3
	Mean	−2	−2.5	−1	0.5	4
	Std error	1.4				
Phase change (degrees)	Replication 1	61	91	85	115	89
	Replication 2	52	72	102	90	79
	Mean	56.5	81.5	93.5	102.5	84
	Std error	8.5				

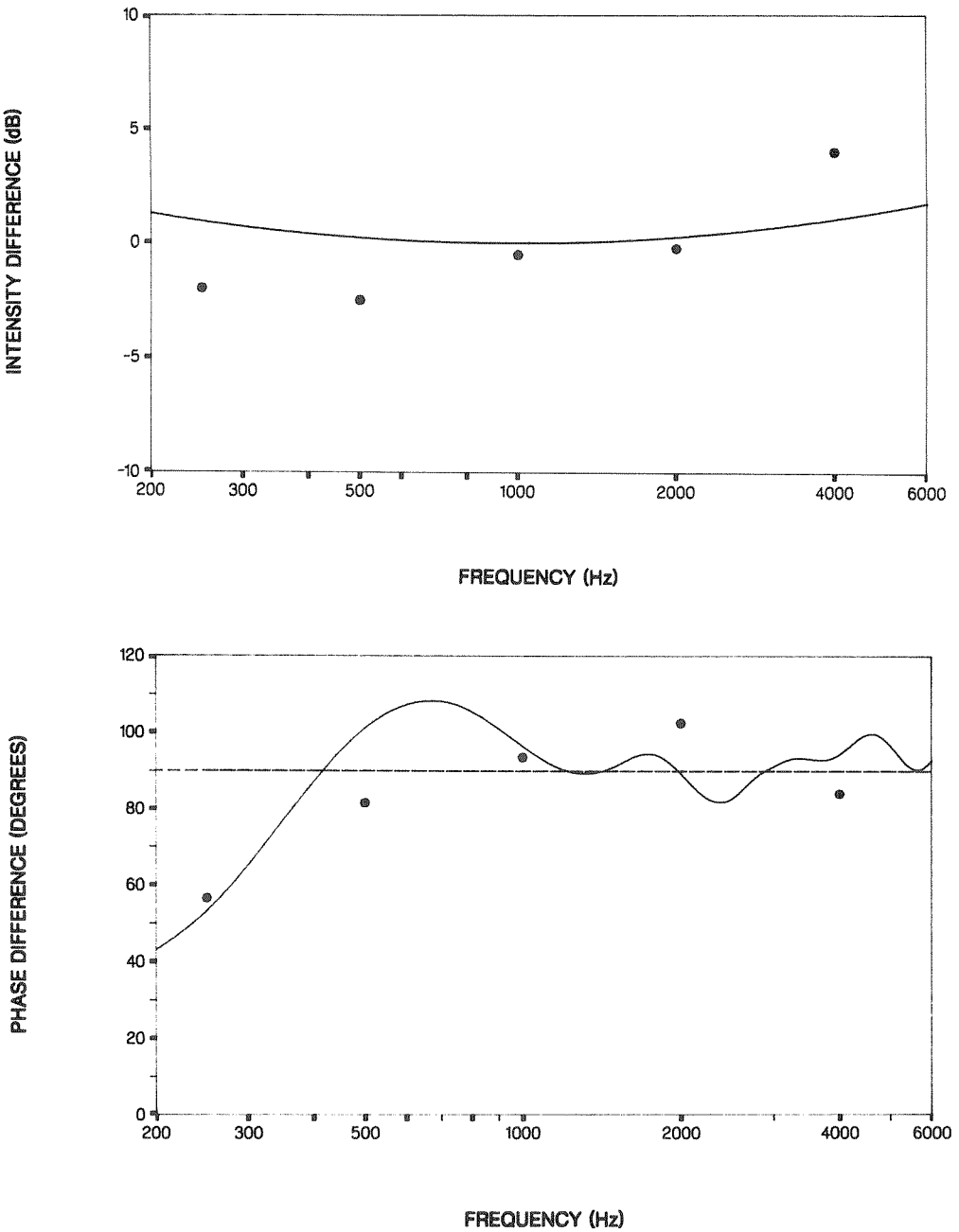

Figure 2.
Amplitude and phase calibration of a digital hearing aid. Measured changes in the amplitude and phase characteristics of a digital hearing aid programmed to produce a 90-degree phase shift are shown. The upper half of the figure shows intensity ratio (in dB), the lower half shows phase shift (in degrees). The plotted points show the data obtained with the cancellation technique; the solid curves show the changes measured electrically.

CALIBRATION OF NONCONVENTIONAL TRANSDUCERS

Several devices have been developed recently that use novel forms of auditory stimulation. In one such system, the Tonndorf Audiometer (13), the ear is stimulated by an effect similar to that of electrophonic hearing (12). Two mylar-coated electrodes are used, one is placed near the ear (e.g., on the mastoid), the other is placed elsewhere on the body (e.g., the arm, or the opposite mastoid). A radio-frequency electrical signal (approximately 60 kHz), modulated by an audio-frequency test signal, is

applied to the electrodes. The electrical field created by the capacitive coupling between the two electrodes produces electromechanical field forces acting on that portion of the body lying between the two electrodes. By a process that is not yet understood, the high-frequency carrier is demodulated and the audio-frequency test signal is transmitted to the cochlea by means of bone conduction.

Another device using nonconventional means of auditory stimulation is the Audiant Bone Conductor (4). In this device, a small magnet housed in a titanium disk is attached to an orthopedic screw and is implanted in the temporal bone. The implanted screw-magnet assembly is caused to vibrate by an electromagnetic coupling placed externally on the skin directly opposite the implanted magnet. The external magnet is held in place by the magnetic coupling between the internal (implanted) and external units. Amplified audio signals transmitted transcutaneously by the electromagnetic coupling cause the temporal bone to vibrate resulting in the transmission of sound to the cochlea by means of bone conduction. The system is currently being used as an implantable bone-conduction hearing device for persons with middle-ear impairments.

The use of nonconventional transducers in audiometers, hearing aids, and other such devices raises an important practical question. How does one measure and calibrate the audio signals generated by these devices? In the two devices described above, the last stage of the transmission path involves bone conduction and, consequently, these signals can be cancelled by acoustic signals reaching the cochlea by another path; e.g., by airborne sound delivered through a standard audiometric headphone or by bone-conducted signals delivered by means of a standard bone-conduction vibrator. The cancellation technique can thus be used to calibrate signals generated by nonconventional transducers, provided those signals reach the cochlea by acoustomechanical means.

Table 2.
Equivalent coupler level for 2000 Hz tone at 120 electrical units.

Subject	Voltage across headphone (millivolts)	Equivalent coupler level (dB SPL)
RD	2.0	63.0
KC	2.5	64.9
HL	2.2	63.8
	Mean	63.9

In order to illustrate this application of the cancellation technique, a calibration check was performed on a Tonndorf high-frequency audiometer. The equipment used was virtually identical to that shown in **Figure 1**, except that the bone conductor was replaced by the Tonndorf audiometer. The electrodes were placed symmetrically, one on each mastoid. A standard audiometric earphone (TDH-39) was used to deliver the airborne signal. The amplitude and phase of the acoustic signal required to cancel the electrical signal were then measured. The amplitude measurement was obtained by first measuring the voltage across the earphone at cancellation and then measuring the sound pressure level generated in a standard 6 cc coupler by the earphone when driven by this voltage. The sound level thus determined is the equivalent sound pressure level, as calibrated in a standard coupler, required to cancel the electrically generated sound.

Table 2 shows data obtained for 3 subjects for a 2000 Hz tone. The electrical signal was set to 120 units as read off the instrument dial. According to the instrument's specifications, this electrical signal should correspond to an acoustic signal of just under 60 dB SPL delivered by a standard audiometric earphone. The data show that this particular instrument is roughly 4 dB out of calibration.

Table 3 shows data comparing equivalent acoust-

Table 3.
Equivalent coupler level as a function of electrical level (Subject HL: Frequency 2000 Hz).

Electrical units	100	105	110	115	120
Measured level (dB SPL)	61	62	62	64	65
Calibrated level (dB SPL)	57.5	58.1	58.7	59.2	59.8
Measured—calibrated levels (dB)	3.5	3.9	3.3	4.8	5.2

Table 4.
Comparison between TDH-39 earphone and ER-3A Tubephone.

Frequency	250	500	1000	2000
Intensity Difference (dB)	0.8	0.6	−1.2	−4.2
	0.9	−1.1	0.7	−3.4
	−0.8	−0.4	−0.5	−4.9
Mean	0.3	−0.3	−0.3	−4.2
Std. dev.	1.0	0.9	1.0	0.8
St. error of mean	0.6	0.5	0.6	0.5
Phase Difference (degrees)	166	−149	−3	−137
	172	−145	5	−120
	160	−140	−8	−133
Mean	166	−145	−2	−130
Std. dev.	6	5	7	9
Std. error of mean	3	3	4	5

ical levels as a function of electrical signal level. The output of the Tonndorf audiometer was increased in steps of 5 electrical units from 100 to 120 units. The measured equivalent sound pressure levels and the nominal calibrated sound pressure levels are shown in the second and third rows of the table, respectively. As before, this particular instrument is seen to be roughly 4 dB out of calibration.

The standard error of the equivalent sound pressure level measurements was found to be on the order of 2 dB. Test-retest variability of successive measurements in which the electrode and earphone remained fixed in position between trials (i.e., neither of the transducers was removed and replaced between trials) was found to be less than 1 dB. The cancellation technique also generated phase data, but those data are not reported since they could not be interpreted meaningfully without a reference phase condition. One possibility is to use a third signal source as a reference (e.g., a signal delivered by a standard bone-conduction receiver), but this was not done since calibration of phase was not of interest in this experiment.

A new transducer system that has found several useful applications in audiology is the Tubephone® developed by Etymotic Research. The Tubephone consists of a receiver coupled to a sound tube terminating in a disposable foam tip that fits snugly in the ear canal. The sound tube is 25 cm in length and is designed so that when the foam tip is inserted into the ear canal, a prescribed frequency response is obtained at the eardrum; e.g., a flat frequency response or a response approximating that of the standard TDH-39 earphone can be obtained, depending on which Tubephone is used.

A model ER-3A Tubephone was calibrated using the cancellation technique. This particular unit has a frequency response approximating that of a TDH-39 earphone. The Tubephone and earphone were each calibrated against a standard bone-conductor receiver (Radio Ear B-70A). These measurements were obtained in pairs at each test frequency, with the bone conductor fixed in position so as to ensure that the signal delivered by the bone-conduction receiver was identical for both Tubephone and earphone. The order in which acoustic cancellation was obtained for the Tubephone and earphone was randomized within each pair of measurements.

The differences in calibration between the ER-3A Tubephone and TDH-39 earphone are shown in **Table 4.** The difference in acoustic output was less than 0.5 dB for frequencies up to 2000 Hz. At 2000 Hz, a difference of roughly 4 dB was obtained. This difference was within the specifications of the two transducers allowing for the observed standard error of measurement. (Note that the standard error of the mean for the intensity data was roughly 0.5 dB and that the standard error of the difference between two means was thus on the order of $\sqrt{2} \times 0.5 =$ 0.7 dB). Frequency response measurements on a standard coupler showed a small drop, approaching 2 dB at 2000 Hz, in the output of the TDH-39 earphone used in this experiment. The Tubephone showed a correspondingly small increase in output in this frequency region.

The data in the lower half of the table show the

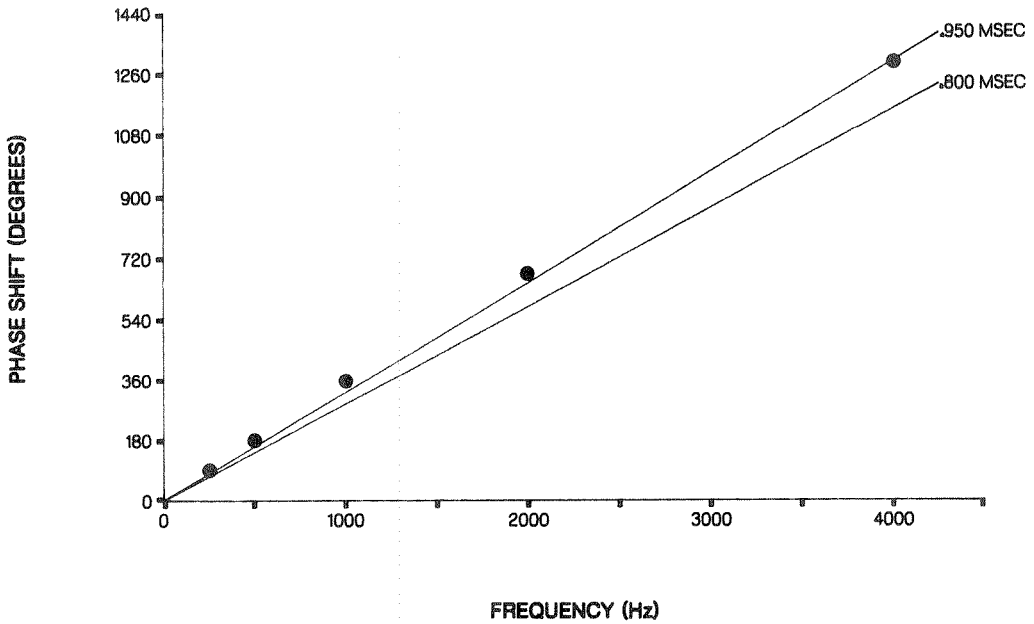

Figure 3.
Comparison between the phase characteristics of a TDH-39 earphone and an ER-3A Tubephone. The data show the measured phase differences after correction for delays in excess of one or more periods. The lower line shows the phase-frequency characteristic corresponding to a delay of 0.8 milliseconds, the nominal delay introduced by the acoustic tube of the Tubephone. The upper line has been fitted to the data by the method of least squares (subject to the constraint that the line pass through the origin) and corresponds to a delay of 0.95 milliseconds.

differences in phase between the acoustic signals delivered by the two transducers. Since the phase characteristics of these two transducers are not specified, it is not known whether or not the observed phase differences are within specifications. It is important to note that the Tubephone has a relative long sound tube, which introduces an acoustic delay of roughly 800 microseconds. At 1250 Hz this delay corresponds to one full period of the signal; i.e., a phase shift of 360 degrees. Since the phase measurements are restricted to a range of 360 degrees it is necessary to correct the measurements for delays in excess of one or more periods of the signal. **Figure 3** shows the difference in phase allowing for this effect. The two sloping lines show phase shift as a function of frequency for a fixed delay. The lower line shows the phase shifts produced by delay of 800 microseconds. The upper line shows the least squares fit to the measured phase differences and corresponds to a delay of 950 microseconds.

CONCLUDING COMMENTS

Although the cancellation technique has been used in research studies for some time (1,10), this technique has not been used very much in audiological measurement. It was only recently, for example, that Kapteyn (5), Khanna, Tonndorf, and Queller (6), and others (3) applied this technique to the difficult problem of calibrating bone-conduction receivers. The measurement procedure described in this paper represents an extension of earlier work of Kapteyn in that the principles of the functional-gain technique have been combined with the cancellation technique in order to measure both the amplitude and the phase characteristics of modern hearing aids and nonconventional transducers. A very useful feature of the cancellation technique in this application is that both intensity and phase characteristics are measured relative to a standard audiometric earphone calibrated in a standard coupler.

The cancellation technique is, in essence, a null method of measurement and as such it is a relatively sensitive technique. In most direct methods of measurement, error of measurement is usually proportional to the magnitude of the quantity being measured. For example, a voltmeter will typically measure a voltage to within X percent of the true voltage, where X is a small percentage. As a consequence,

the absolute magnitude of the measurement error increases with signal level. In the null method of measurement, measurement error is typically proportional to the magnitude of the difference term which is usually very small in comparison with the signals being cancelled. As a result, the absolute magnitude of the measurement error in the null method is usually substantially less than that of direct measurement.

A useful additional feature of the cancellation technique is that the test and reference signals can be relatively high in level so that changes in the functional gain of a hearing aid (or differences in functional gain between a test and reference hearing aid) can be measured at levels typical of everyday use. In the usual method of measuring functional gain, the test signals are at threshold level and may be masked by the internal noise of the hearing aid. Many hearing aids also have a frequency response that varies with signal level, and for those instruments functional gain should be measured at normal operating levels.

The cancellation technique can also be used to measure the audibility of the harmonic distortion generated by a hearing aid at typical operating levels. In using the cancellation technique, only the fundamental frequency component of the test and reference signals are adjusted to cancel each other. If harmonic components are present (e.g., due to nonlinear distortion in the hearing aid or transducer being calibrated) those distortion components will be audible. The levels of the test and reference signals can thus be adjusted to find the level at which harmonic distortion is just audible. This provides a direct measure of the audibility of harmonic distortion components in hearing aids and transducers in general.

A problem in using a bone-conduction receiver to provide the reference signal is that these units are usually limited in their frequency response. In **Table 1**, for example, data were not obtained for frequencies above 4000 Hz because of bandwidth limitation of the bone-conduction receiver. It is possible to use a Tonndorf audiometer to generate the reference signal, in which case measurements over the entire audio-frequency range can be obtained. In addition to the very wide frequency range of the bone-conducted signals generated by this instrument, the precision of electrode placement can be very high, thereby reducing the problem of intertest variability. A serious practical problem with conventional audiometric earphones at high frequencies is that of standing waves in the ear canal. Standing waves at high frequencies can produce substantial changes in sound pressure level at the eardrum resulting from small changes in earphone placement.

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