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NEW KNOWLEDGE ABOUT THE FUNCTION OF THE HUMAN MIDDLE EAR: DEVELOPMENT OF AN IMPROVED ANALOG MODEL

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ABSTRACT

Conventional teaching regarding the acoustic function of the human middle ear is that it serves as an impedance matching system to offset the loss that occurs when sound passes from the low-impedance sound field to the high-impedance cochlear fluid. A transformer analogy is often used with the pressure transformation produced by the eardrum; footplate area ratio and the lever ratio considered to be approximately 27 dB. Recent data on middle ear function has shown this to be only partially correct. A transformer analogy is not appropriate since the pressure gain of the middle ear decreases above 1000 Hz and does not depend on the cochlear load at all frequencies. Experiments are described of umbo, malleus short process, and stapes displacement in human temporal bones using a laser Doppler measuring system (LDS). The measurements support previous studies that indicate that in addition to a roll-off in tympanic membrane function above 1000 Hz, there is slippage in the ossicular lever system that causes an increasing "lever ratio" above 1000 Hz, thought to be caused by translational movement of the ossicular rotation axis near the short process. An improved analog circuit model of the external and middle ear has been developed that produces results equivalent to those found in the temporal bones.

The acoustic function of the middle ear has been described in many publications and its basic principles of sound transduction are well known. Much of this knowledge has been obtained from measurements of ossicular and tympanic membrane vibration in animals, particularly the cat. Multifrequency impedance measurements at the tympanic membrane (TM) have also provided useful information. The relatively new development of sophisticated noncontacting measuring systems capable of determining displacement in the submicron range have provided a means to study the human middle ear in greater detail, in both the cadaver and live humans. Several recent studies have shown that the human middle ear has certain differences in function compared to animal ears and that measurements in fresh cadavers are quite similar to those in the live human. In addition, data have become available to question some of the classic thinking about the middle ear, such as the theoretic loss produced when sound passes from air into the cochlear fluids, the mechanical transformer analogy, the amount of pressure gain provided by the middle ear, and the constancy of the lever ratio.

This paper reviews selected aspects of human middle ear acoustic function in which the concepts are changing and describes a new analog middle ear network model. Emphasis is on clinical applications of this information.

Recent studies, performed in fresh human temporal bones and live humans,[1–4] on the middle ear transfer function suggest that the human middle ear becomes increasingly inefficient above 1000 Hz, not 2000 Hz as described by Békésy.[5] In this respect the human middle ear transfer function is different from that of the cat and guinea pig whose high frequency roll-off begins at a higher frequency (1500 Hz for cat, 4000 Hz for guinea pig) and is not nearly as rapid.[6–8]

Three causes for this inefficiency have been identified. (1) Decrease of TM function at higher frequencies attributable to the TM breaking up into

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smaller vibrating portions rather than a piston-like motion.[9] (2) Inefficiency in lever function above 1000 Hz.[1,10] Part of the high frequency roll-off in the human middle ear appears to be related to slippage in the ossicular transmission system. A constant "lever ratio" in the human of 1.3:1 is regularly quoted, based on anatomic dimensions and measurements of umbo and stapes head vibration at low frequencies or in temporal bones with the cochlea drained.[11-13] The ratio of umbo to stapes motion increases when measured at higher frequencies in intact bones. The transmission loss between the umbo and stapes head appears to be attributable to translational movement in the axis of rotation of the ossicles above 1000 Hz. The reason for this is unclear; it may be an age effect, attributable to laxity in the suspensory ligaments or the TM, or a species effect, since in the cat and guinea pig the ratios are relatively constant up to high frequencies.[6,8] This slippage produces an additional 4–6 dB per octave decrease in stapes displacement between 1000 and 6000 Hz. (3) The 3000-Hz notch in human middle ear sound transmission,[14,15] similar to the 4000 Hz notch in the cat,[6,7,16] is produced by a mastoid cavity and aditus ad antrum resonance that decreases the effective sound pressure difference across the TM. It can be eliminated by blocking the aditus.[15] Thus, the 1000–3000 Hz region in the human appears to be the steepest part of the higher frequency roll-off, which averages 12–15 dB per octave and is obtained by measuring stapes displacement for a constant sound pressure level at the TM.[17] Whereas ossicular displacement and velocity measurements have been used to support the contention that the human middle ear is less efficient above 1000 Hz, sound power measurements in and out of the middle ear provide similar results.[18]

Significant individual variation, up to 25 dB, exists in middle ear function at key hearing thresholds in both animal and human experiments. Dallos[19] and Fleischer[20] have stated that the external and middle ear transfer function determines the threshold hearing curve. The authors have previously suggested that the variations in external and middle ear function between individuals explains a significant part of the 25-dB range in hearing thresholds that occur in young subjects with normal ears, particularly at 1000–4000 Hz.[21,22]

It appears likely that variations in cochlear mechanics may also contribute to the variations in middle ear function at some mid frequencies. Studies in our laboratory in five fresh human temporal bones revealed that the human cochlea–stapes input impedance is relatively flat from 2000 to 6000 Hz, with a value of about 0.5 million acoustic ohms (cgs).[23] Below 1400 Hz the impedance was lower, with a mean of 0.24 million ohms rising to the higher figure between 1400 and 2000 Hz. In all five bones, a notch in the impedance was present near 1000 Hz. Removal of the cochlear load does not change umbo displacement above 1000 Hz.[1,16]

Postmortem artifacts are not thought to be the cause of the poor high-frequency function. There is substantial evidence that human temporal bones kept moist and used within a week after death maintain their premortem middle ear mechanical function as long as drying does not occur.[1-4,14,15] Impedance measurements made in fresh temporal bones are the same as in live humans and umbo displacement in response to a constant sound input is similar in live ears and temporal bones.[24,25] Abnormal values have been reported in postmortem specimens. The high middle ear impedance found at low frequencies by Zwislocki and Feldman[26] was undoubtedly produced by the negative middle ear pressure that occurs after death.[27] Onchi's temporal bone impedance data showed an increase in reactance at low frequencies when compared to normal subjects; no differences were present above 1000 Hz.[28] The reason for this is not known, but the method he used to measure impedance was difficult and subject to experimental error. The middle ear muscles are nonfunctional in the cadaver, but their function has been well studied in live subjects; muscle inactivation produces essentially no effect on thresholds above 1000 Hz.[29]

MIDDLE EAR ANALOG MODELS

Modeling of middle ear function using a circuit model has been described by Moller,[30] Zwislocki,[31] Shaw and Stinson,[32] Lynch,[16] Killion and Clemis,[33] and Kringlebotn,[34] as well as several others. The subject has recently been reviewed by Shera and Zweig.[35] Accurate models could be helpful to the otologist in understanding the effect of various anatomic modifications on middle ear sound transmission, in particular modifications commonly performed during middle ear and mastoid surgery.

In 1981, Killion and Clemis[33] described a circuit model for the external and middle ear based on the Shaw and Stinson[32] dual-piston eardrum version of Zwislocki's[31] middle ear model; the model was termed SKEAR 6 (Shaw, Killion Ear, Version 6). This model was based on a "good young ear" and did not show the 3000 Hz notch in transmission nor the change in the lever ratio with frequency found in the majority of human temporal bones.

A comparison between the measured average umbo and stapes displacement obtained by Gyo, et al and the calculated umbo and stapes displacement based on the SKEAR 6 model showed excellent agreement between the measured and calculated umbo displacement at all frequencies; however, the agreement between measured and calculated stapes displacement was excellent only below 500 Hz and relatively poor above that frequency.[1,33] A model consistent with the findings in an average ear was needed; an initial report on that model has been previously presented.[36]

MATERIAL AND METHODS

Umbo, Short Process, and Stapes Displacement Measurement in Human Temporal Bones

Twenty human cadaver temporal bone specimens were used for the study. Specimens were obtained within 48 hours after death using a Schuknecht oscillating bone saw. The age distribution was 53 to 87 years with an average age of 68.9 years; all were male. The temporal bones were preserved in 1 to 10,000 Merthiolate solution at 5°C until used; all experiments were performed within 6 days after death.

After confirming by visual inspection that the TM was intact and the middle ear normal, the TM was coated with a drop of safflower oil. Any openings in the mastoid cells to the outside were covered with bone wax and molding clay and the bone was wrapped in a rubber finger cot to prevent drying. Dental cement was applied to make a block to hold the bone in a temporal bone holder. The bony wall of the external auditory canal (EAC) was carefully drilled down to 3 mm from the tympanic membrane annulus. Ten bones were used for umbo and short process measurements. A 1 × 1 mm area of the bony surface at the umbo and short process was exposed using an argon laser to allow the attachment of a reflective target. A 1 × 1 mm piece of reflective Mylar tape, weight 0.1 mg, was attached to the short process and the umbo using cyanoacrylate cement (Krazy Glue).

Ten additional bones were used for umbo and stapes footplate displacement measurements. Measurement of stapes displacement required a simple mastoidectomy with wide opening of the facial recess. The intratympanic portion of the facial nerve was removed to provide visualization of the footplate. A 1 × 1 mm piece of reflective tape, weight 0.1 mg, was placed on the center of the footplate and held with cyanoacrylate cement. A clear glass coverslip was placed over the mastoidectomy opening, and all other mastoid air cell openings were closed with bone wax and modeling clay. The prepared specimen was coated with dental cement, placed in a temporal bone holder and fixed onto a vibration isolation table; a modified aural speculum was placed in the EAC and adjusted so that the entire TM could be viewed from the EAC. The speculum had two small channels, one for sound input and the other for a probe microphone to monitor and control the sound pressure level at the TM. The outer opening of the speculum was covered with a glass coverslip to produce a closed system.

The laser Doppler vibrometer (DISA 9055) measures the velocity of a vibrating surface by utilizing a helium-neon laser as a light source. The laser beam is split into two paths; one beam is frequency shifted by 40 MHz relative to the other by a frequency shifter (DISA 55NI2). The frequency shifted beam is focused through the speculum onto the reflective target by means of the front lens on the laser, the other frequency unshifted beam is used as a reference beam. The reflected beam from the target is received by an optical heterodyne detector, which consists of two photo-diodes. The two signals are mixed with a 40 MHz isolator signal in the frequency shifter and the different signal fed to the Doppler signal processor in the frequency tracker unit (DISA 55N21). The output signal is a voltage proportional to the velocity of the target. The method used is similar to that described by Bunnen and Vlaming,[7] Vlaming and Feenstra,[4] and Goode et al.[25]

The pure-tone sound stimulus was delivered to the TM through one side channel of the speculum in the ear canal. A University (Sylmar, California) speaker driver ID-40T driven by a Breul and Kjaer (Marlborough, Massachusetts) signal generator (B&K 1023) provided the sound input. The sound pressure level (SPL) at the TM was monitored by a probe microphone (7C Etymotic Research, Elk Grove Village, Illinois) inserted into the second channel of the speculum. The SPL at the TM was 104, 114, and 124 dB for the umbo and stapes measurements and 94, 104, 114, and 124 dB for the umbo and short process measurements. Twenty-one pure-tone frequencies between 400 and 6000 Hz were used for the umbo and stapes measurements and 22 frequencies between 200 and 6000 Hz for the umbo and short process measurements. The SPL at the TM was held constant by the probe microphone whose output was amplified by a B&K 2603 amplifier and then returned to the signal generator. The voltage output from the laser Doppler measuring system (LDS) and the stimulus sound signal were monitored on a dual channel oscilloscope and recorded on an X-Y recorder (B&K 2308). The voltage output is proportional to the velocity of the target. Voltage at each frequency was converted to peak-to-peak displacement in microns (μM) using the formula:

$$\text{Displacement } (\mu\text{M}) = \frac{\text{Vp} - \text{p}}{2\pi\,\text{F}} \times \text{C}$$

where Vp–p = peak-to-peak voltage
C = calibration factor of the LDS, 31.64×10^{-3} m/sec/volt
F = frequency in kHz
$\pi = 3.14$

Whereas stapes volume velocity and not displacement is the measurement that best correlates with the hearing threshold curve, peak-to-peak displacement is conventionally used to report ossicular vibration data.

In seven temporal bones, the phase of the velocity at the umbo, short process, and stapes footplate was compared with the phase of the input sound signal using the dual channel oscilloscope and a digital phase meter (Model 6200A Krohn-Hite, Avon, Massachusetts). The phases were then compared to each other by subtracting the phase of the umbo from that of the short process and the stapes footplate. Phase measurements were made at 104 and 114 dB SPL.

RESULTS

Umbo and Stapes Footplate Displacement

Umbo and stapes footplate displacement at 104 and 124 dB SPL inputs are shown in Figure 1. There was little evidence of nonlinearity. The lever ratio at 114 dB SPL at the TM is shown in Figure 2. The data shown in Figure 1 reveal that mean umbo displacement rolled-off at –10.4 dB per octave from 1000 to 4000 Hz, whereas stapes displacement rolled off at –13.4 dB per octave. The lever ratio had a mean value of 2.2 below 1400 Hz, rising to 5.2 at 4000 Hz, and reaching a peak of 6.1 at 6000 Hz.

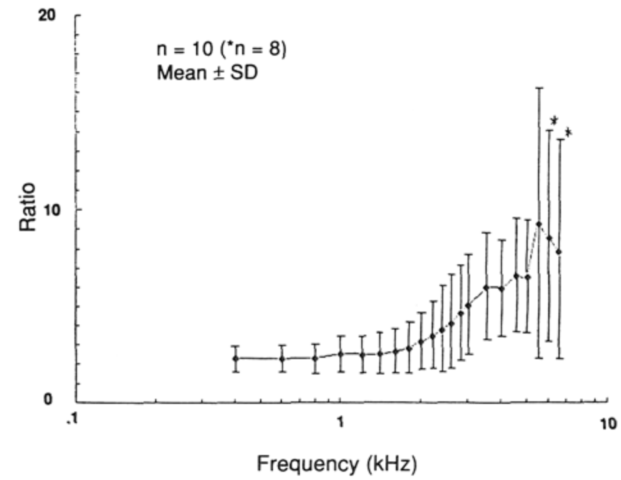

Figure 2. The lever ratio (umbo displacement:stapes footplate displacement) produced by 114 dB SPL at the TM in the same ten temporal bones assessed in Figure 1.

Umbo and Short Process Displacement

Umbo and short process displacement in ten temporal bones are shown in Figure 3; umbo and short process displacement in a representative ear is presented in Figure 4. Umbo displacement was greater than the short process displacement at all frequencies from 200 to 6000 Hz. Above 800–1000 Hz, umbo displacement begins to decrease at a rate of approximately –8.0 dB per octave. The short process resonates at around 1600 Hz. Above this frequency, the decrease in short process displacement is about the same as that of the umbo. At lower frequencies below 1600 Hz, short process displacement is about 12 dB lower than that of the umbo.

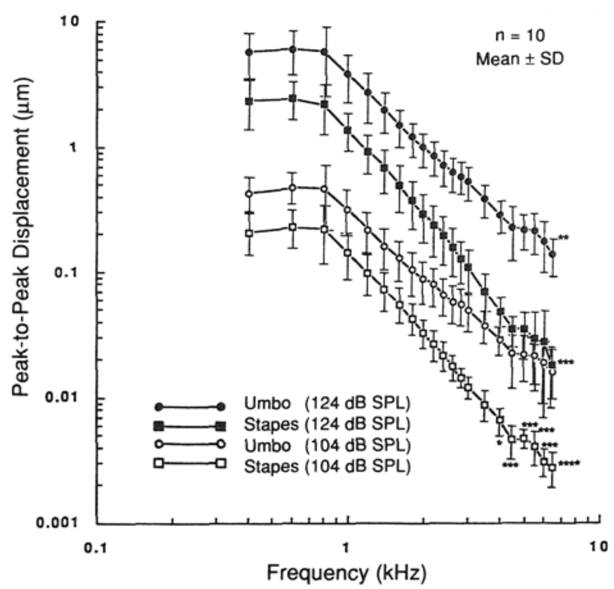

Figure 1. Umbo and stapes footplate displacement in ten human temporal bones at 104 and 124 dB SPL inputs at the TM. (Mean age 72.0 years, range 65–80 years). (* = 9 ears; ** = 8 ears; *** = 7 ears; **** = 6 ears)

Relative Phase of Umbo and Stapes Displacement

Comparative phase measurements of umbo and stapes displacement showed the stapes to lag behind the umbo by –6 to –29 degrees below 1000 Hz increasing gradually to –61 to –96 degrees at 2000 to 5000 Hz. Short process displacement also lagged behind the umbo displacement by –55 to –105 degrees below 1500 Hz, crossing the zero line at 2500–3000 Hz and increasing relatively rapidly to +80 degrees near 4000 Hz.

Revision of the SKEAR 6 Model

In the revised model, several changes were made. The original Zwislocki model values for the middle ear cavities were used, as marked on the analog circuit shown in Figure 5, as the SKEAR 6/Z modification. This change was made based on the finding that the notch in the umbo displacement curve near 3000 Hz was a typical finding in temporal bones attributable to the increase in middle ear cavity pressure caused by the resonance with the mastoid air cells through the aditus.[1,9,10] This notch appears in the calculations, if the original Zwislocki values are used.[31]

The next revision was to place 60 mg at Lmw, between two shunt compliances of 0.44 μF each. The first compliance (Cs) with its accompanying resistance (Rs) originally represented the "squishiness" of the incudostapedial joint. A second compliance (Cmw) was added to the model to allow malleus-head axis laxity or "wobble." With this modification, the agreement between calculated and measured umbo *and* stapes displacement becomes acceptable (Fig. 6).

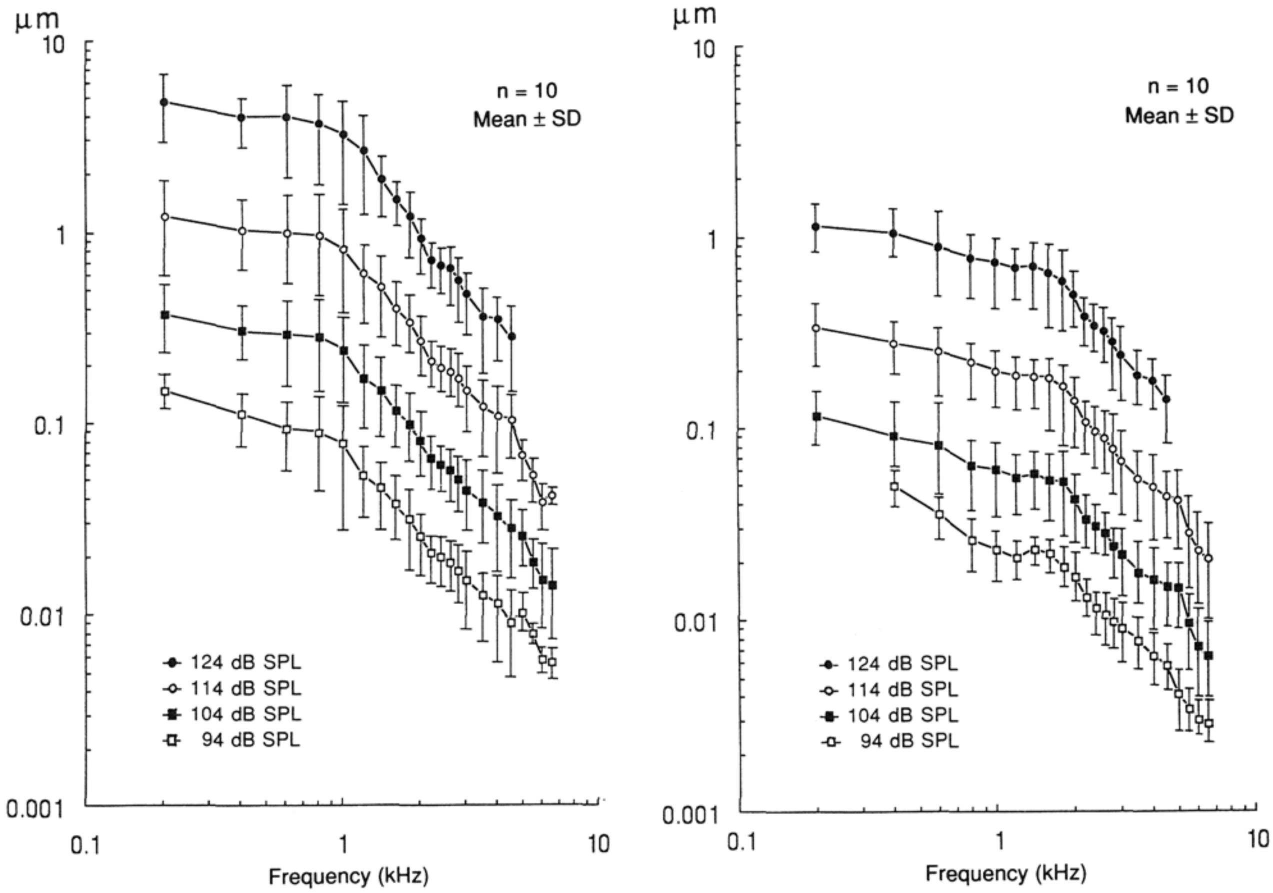

Figure 3. *Left,* Umbo displacement in ten temporal bones at 94, 104, 114, and 124 dB SPL inputs at the TM. (Mean age 65.8 years, range 53–87 years). *Right,* Short process displacement in the same ten temporal bones at 94, 104, 114, and 124 dB SPL inputs at the TM.

The final issue was the proper value for the cochlear input impedance. The authors had previously obtained a result of roughly 1 million acoustic ohms (cgs) measured in one temporal bone and believed that measurement should be used rather than the 365,000 ohms suggested by Zwislocki and used in SKEAR 6.[31] This is Rc in the model. Increasing Rc from 365,000 to 1,000,000 ohms in the original SKEAR 6 model also increased the calculated impedance seen looking into the eardrum, an increase that was unattractive because SKEAR 6 had modeled measured real-ear impedance data so well. Increasing Rc to 1 Meg (SKEAR 7C modification in Fig. 4) with the addition of the malleus axis wobble section produced a net result close to that of the original SKEAR 6 eardrum impedance. Further testing in five temporal bones with the improved method has shown that the stapes–cochlea input impedance in human temporal bones is about 0.5 million acoustic ohms at higher frequencies.[23] Figure 7 shows the high-frequency loss calculated with the revised model, called SKEAR 7Z, compared to the presumed "good young ear" modeled in the original SKEAR 6 version.

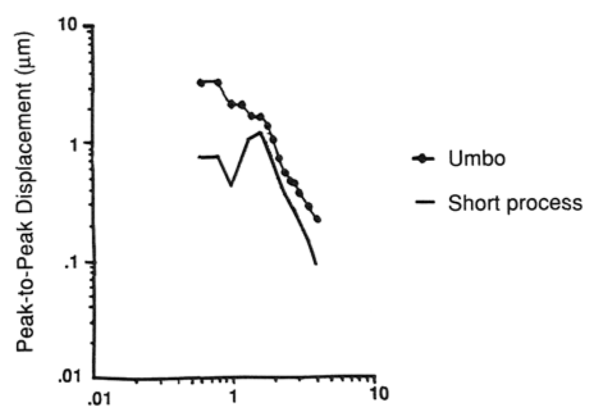

Figure 4. Umbo and short process displacement in one representative temporal bone at 124 dB SPL input at the TM.

In the future, finite element models will, in all likelihood, be the best type of model for the middle ear since they provide more information than the circuit models.[37]

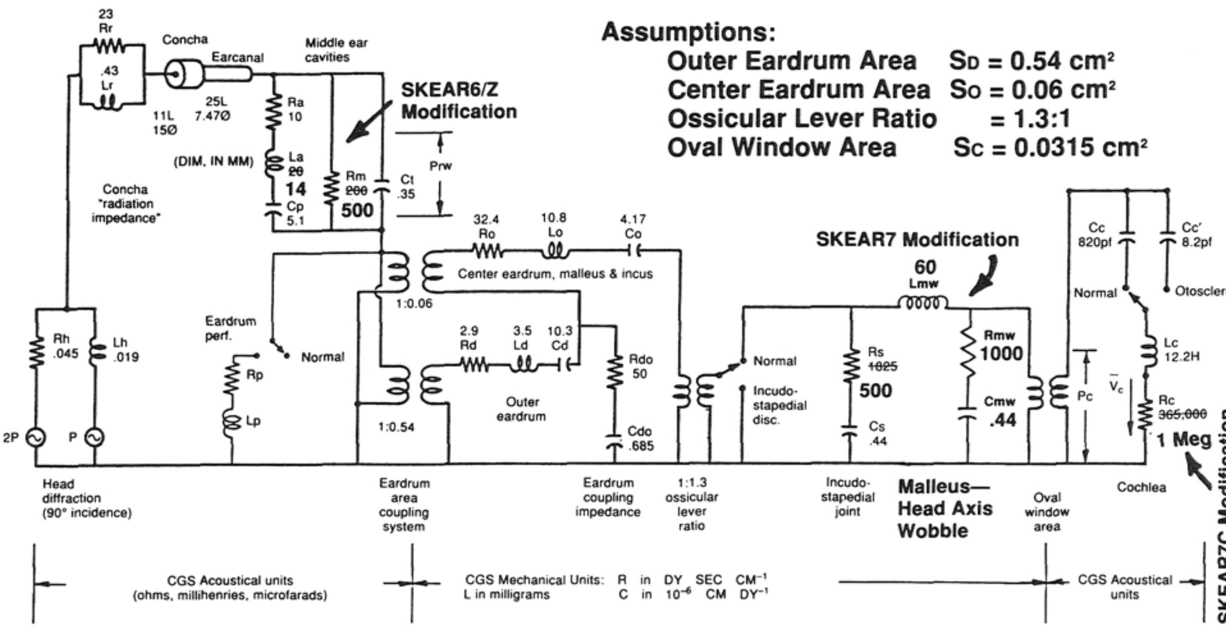

Figure 5. Modifications of the original Killion and Clemis analog model (SKEAR 6) are shown and include SKEAR 6/Z, SKEAR 7, SKEAR 7C, and a modification for malleus head-axis wobble.

DISCUSSION

The Transformer Analogy

A common description of middle ear function is that it acts as a mechanical transformer matching the low impedance of sound in the air of the ear canal to the relatively high impedance of the cochlear fluids. This analogy is incomplete and, despite its popularity, should be revised, for several reasons. First, the classic audio transformer is becoming a device of the past, being replaced by modern semiconductor technology and thus will have less and less meaning in the future as a concept. Second, the transformer analogy implies a fixed ratio of pressure gain over the frequencies of interest, usually 200–8000 Hz. This is definitely not the case, as has been described. The human middle ear pressure gain varies with frequency; the major gain occurs in the lower frequencies with a peak near 1000 Hz and then rolls off above that frequency. This is shown for stapes displacement in Figure 1. Whereas the transformer model may be suitable for low frequencies, a more complex circuit model, such as the model described in this paper, is necessary to understand function at higher frequencies above 1000 Hz: Third, Lynch found that the

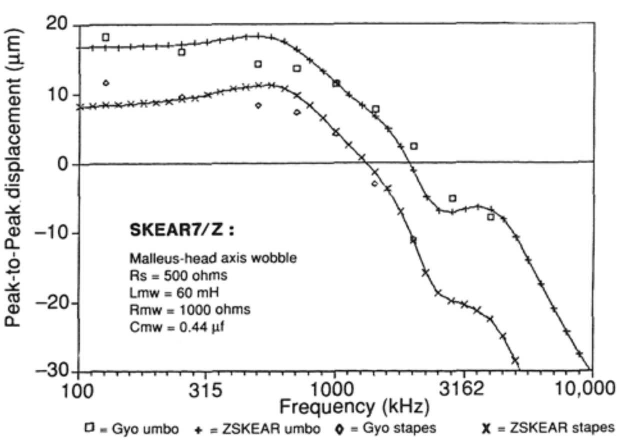

Figure 6. Comparison of measured mean umbo and stapes displacement obtained by Gyo et al[1] and calculated values using the Killion and Clemis (SKEAR 6/Z) modified analog ear model.

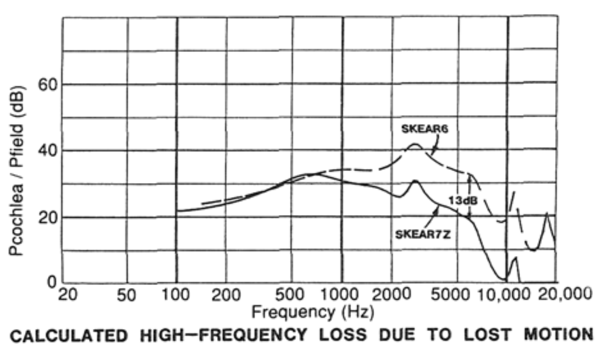

Figure 7. Comparison of the sound pressure gain produced by the original Killion and Clemis analog model (SKEAR 6) and the final modified model (SKEAR 7/Z), which includes the modification for malleus head-axis wobble.

middle ear admittance measured at the TM is independent of the cochlear input admittance at most of the key hearing frequencies and that the admittance ratio between the TM and the cochlea is not the square of the pressure ratio, as would be expected with a transformer analogy.[16]

Impedance Matching by the Middle Ear

Killion and Dallos have suggested that the best formula to quantitate the pressure transformation required to produce a perfect impedance match between the sound field and cochlea is provided by the formula:

$$\frac{3000}{\sqrt{F}}$$

where F = frequency in Hertz.[38]

Using this formula, it can be seen that the impedance mismatch that must be overcome by the external and middle ear is not fixed at 30 dB, a number still provided in many textbooks, but decreases linearly with a rise in frequency. At 100 Hz, the ratio is 300 or approximately 50 dB, whereas at 2500 Hz the ratio is 60 or about 36 dB. Therefore, less pressure gain is required at higher frequencies to correct the mismatch. The 30-dB figure is based on the 99.9 percent reflection of sound when passing from air into seawater; this concept is not correct for the small structures of the ear.

Lever Ratio

The temporal bone experiments reveal a difference in the slopes of umbo and stapes vibration above 1000 Hz, so that an increase in the lever ratio occurs with rising frequency as shown in Figures 1 and 2. These results are similar to those previously reported, using a less sensitive measuring system, and reveal that the human lever ratio is not fixed, but rises with frequency above 1000 Hz.[1] It appears that slippage occurs in the ossicular system at higher frequencies. The other explanation of an increase in force with frequency does not appear possible. The location of the slippage was not initially apparent. Measurements of displacement of the umbo, incus lenticular process, and head of the stapes in 14 temporal bones found only a relatively constant 2-dB difference between the stapes head and the lenticular process up to 2000 Hz.[1] In the human, the incudomalleolar (I-M) joint is fixed at physiologic sound pressures but does bend at high static pressures.[10,39] Gluing the I-M joint with cyanoacrylate glue in one bone did not change the displacement amplitudes.[1]

Slippage is believed to be attributable to a relative laxity in the ossicular rotation axis, which lies on a line between the anterior malleolar ligament and the posterior incudal ligament close to the short process of the malleus. This axis is not fixed, but moves in and out as well as rotating. This movement would tend to decrease the efficiency of the lever action at higher frequencies as well as shift the axis of rotation, both of which have been noted to occur.[10,39] This slippage produces a loss of about –4 to –6 dB per octave above 1000 Hz in displacement between the umbo and stapes, over and above the –9-dB per octave loss in this frequency range produced in part by the loss of the piston action of the TM.[25] Whereas this slippage may be detrimental acoustically, it probably is useful in protecting the cochlea from excessively high pressures. Observations of ossicular motion show that umbo and stapes displacement is not invariably in and out but can be rotational at some higher frequencies.[1,40] This also may contribute to the inefficiency of the lever system.

The 3000 Hz Notch

As previously stated, the slope of umbo and stapes displacement in individual ears is steepest between 1000 and 3000 Hz, due to a notch near 3000 Hz secondary to the resonance that occurs in the middle ear cavity at this frequency. This produces an increase in sound pressure on the middle ear side of the TM, decreasing the net effective sound pressure across the TM that controls its movement. One effect of this notch is to smooth out the soundfield-to-stapes transfer function at the expense of the middle ear transfer function. Blocking the aditus eliminates the notch, producing a 5–6 dB improvement at the resonant frequency.[15] Clinically, this would suggest that a modified radical mastoidectomy, which eliminates the connection of the mastoid to the middle ear, would be better acoustically in this frequency range than an intact canal wall mastoidectomy.[14]

Short Process Displacement

If there is translational movement at the rotational axis of the malleus and incus, the short process should move in and out, since it lies near the axis. Figures 3 and 4 show the in and out displacement of the short process compared to the umbo. Note the relative increase in short process displacement with frequency.

The question of the location of the rotational axis and whether it moves with a change in frequency has been addressed in our laboratory by Aritomo.[40] He did not make measurements at multiple sites on the malleus, but by measuring displacement at the umbo and the short process, knowing the length of the malleus, the distance between the two sites, and the phase difference between the two sites, it was possible to estimate the relative position of the axis of rotation. At low frequencies, the axis is located at the neck of the malleus, and as the frequency increases from 1000 to 2000 Hz, the axis moves upward toward the malleus head. Gundersen used an electromagnetic transducer to compare various vibrating points on the malleus;[10] he found that there were no points where the amplitude of displacement was zero, but the smallest amplitude was obtained near the neck of the malleus. This would imply that the rotational axis for the malleus–incus vibration is at the malleus neck and that translational (in and out) movement is present as well as rotation. He found the rotational axis moved upward with an increase in frequency, particularly above 1000 Hz. Aritomo's data support the findings of Gundersen.[10,39]

The translational displacement of the rotational axis of the malleus to the umbo increases between 1000 to 2200 Hz and then remains constant at higher frequencies. This higher frequency movement produces an apparent loss of efficiency in the transfer of sound vibration between the malleus and stapes above 1000 Hz as described by Gyo et al.[1] The reason is that for a simple lever to function properly, its fulcrum or axis of rotation must be stable. Significant translational movement of the axis would subtract from any theoretic gain in force produced by the lever. As a result, the human ossicular lever ratio increases with frequency, and stapes displacement rolls off more rapidly than umbo displacement above 1000 Hz.[8,12]

There are certain clinical implications of these findings. When an incus replacement prosthesis (IRP) or incus–stapes replacement prosthesis (ISRP) is required, the prosthesis is often placed to the neck of the malleus, because it is a convenient location in regard to the angle between this site and the stapes head or footplate. The best theoretic location on the malleus for a prosthesis would be as close to the umbo as possible, since this site has the largest displacement for a given SPL at the TM. The reason that placement of the prosthesis at the neck produces vibration of the stapes is that there is translational movement of the axis at the neck. Vibration from the malleus neck in response to sound pressure at the tympanic membrane would be expected to be less then at the umbo, particularly below 1000 Hz. Based on this study, the difference should be about 10 dB at low frequencies below 1000 Hz; decrease to around 6 dB at 1000–2200 Hz and remain there above 2200 Hz. Removal of the incus probably increases malleus neck movement by removing some of the suspensory ligaments and incus mass; section of the tensor tympani tendon may also improve neck displacement. In summary, it would be expected that at mid and low frequencies there would be a relative improvement of 6–10 dB by placement of an IRP or ISRP between the umbo and the stapes compared with placement between the malleus neck and stapes, all things being equal.

Placing the head of the prosthesis to the umbo may produce an excessively acute angle between the shaft of the prosthesis and the malleus handle that could be detrimental to sound transmission and possibly predispose to later slippage of the prosthesis. This is particularly true when there is retraction of the tympanic membrane with medial displacement of the umbo. At times, even the neck location produces too acute an angle and substitution of a prosthesis connecting the posterosuperior TM to the stapes head or footplate should be considered.

Mills made measurements of the angle between the malleus handle and stapes head in 31 human temporal bones and 20 patients at the time of otologic surgery. He found the range to be from 14 to 71 degrees (mean 49 degrees) in the temporal bones and from 0 to 79 degrees (mean 63 degrees) in the patients.[41]

As the angle of the IRP or ISRP to the direction of malleus vibration increases, the theoretic sound transfer of the vibration from the umbo to the stapes decreases in proportion to the square of the cosine of the angle the prosthesis makes with the direction of malleus handle movement (angle A in Fig. 8). This analysis assumes that the angle between malleus handle vibration and the IRP or ISRP is identical (or nearly so) with the angle between the prosthesis and the direction of vibration of the stapes head or footplate, and that the malleus handle and stapes both move in the same direction. By vector analysis, there will be a loss in force or decrease in displacement

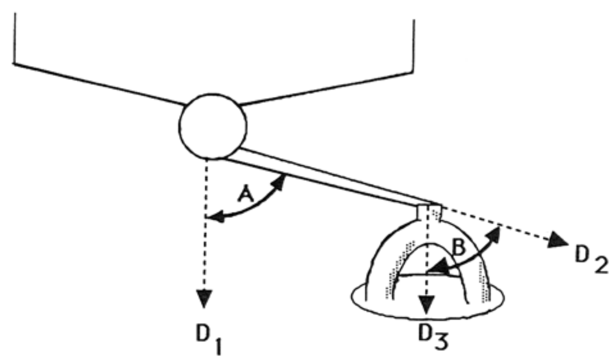

Figure 8. Diagram of the effect of angulation on relative displacement of a prosthesis from the malleus handle to the stapes head. The magnitude of the vectors is not drawn to exact scale. D_1 is the displacement vector of the malleus; angle A is the angle between the prosthesis and the direction of malleus handle vibration. D_2 is the displacement vector of the prosthesis and is equal to the cosine of angle A $\times$ D_1; D_3 is the displacement vector of the stapes, moving in the same direction as the malleus; B is the angle between the prosthesis vector (D_2) and the stapes displacement vector (D_3) and is approximately equal to angle A; D_3 is equal to the cosine of angle B $\times$ D_2; D_3, the stapes displacement, is equal to the cosine of angle A^2 times D_1, the malleus displacement. Whereas a malleus to stapes head prosthesis is considered in the figure, a malleus-to-footplate prosthesis could be analyzed in the same manner.

between the in and out movement of the malleus and the prosthesis equal to the cosine of angle A, and, at the same time, a decrease in displacement between the prosthesis and the stapes head or footplate equal to the cosine of angle B. Since angle A = angle B, the velocity or displacement loss at the stapes equals the cosine of angle A squared times the umbo velocity or displacement. Assuming no friction or slippage, a 45 degree angle would produce a theoretic change in sound transmission between the umbo and stapes of 0.5 or a 6 dB loss; at an angle of 60 degrees, the change is 0.25 or a 12 dB loss. It would appear that an umbo or near umbo location is optimal only if the angle between the malleus handle vibration and shaft of the prosthesis is 45 degrees or less, assuming good fixation of the prosthesis to the malleus can be obtained. If good fixation cannot be obtained at or near the umbo, and the neck allows good fixation and an angle less than 45 degrees, this site should be considered. If the angle is more than 45 degrees at the neck or handle, the posterosuperior TM location would be preferable. Vlaming and Feenstra have also discussed these issues.[42]

Although the title of this paper implies "new" information, this is a relative term. Readers with a detailed knowledge of the middle ear literature may be aware of much of the information provided here. However, an additional group of human temporal bone experiments has been presented that support the findings of previous investigators along with a refinement of an existing external and middle ear analog circuit.

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