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NOTICES

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MATHEMATICAL MODEL OF THE EAR'S RESPONSE TO WEAPONS IMPULSES

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ABSTRACT

We develop a mathematical model of the ear's response to intense sounds (over 120 dB) that starts with pressure in the free-field, accounts for the effects of the head, external ear, middle ear, and cochlea, and ends with predictions of basilar membrane motion. The model's structure parallels the ear's anatomy and function and fits hearing loss data for weapons impulses that are not explainable by any other mechanism. The model suggests that upward movements of the basilar membrane produce the stresses that are the primary cause of damage.

INTRODUCTION

The previous paper in this symposium pointed out deficiencies in present damage-risk criteria (DRCs) which ranged from a lack of theoretical foundation to outright inaccuracy, especially where large caliber weapons are concerned (Price, 1986b). This analysis was based in part on a series of experiments suggesting that the ear's response to intense stimulation can be understood best as a function of mechanical stress within the inner ear (Price, 1979; 1981; 1982; 1983a; 1983b; 1983c; 1986a). However, we can't measure stress within the inner ear directly; therefore mathematical modeling of the ear's behavior at high intensities is the most viable alternative.

Such a model is the subject of this paper. Given that the participants in this symposium have technical backgrounds in areas other than cochlear modeling, we will try to present enough detail to make the implications of the model intelligible. For those interested in the complete development, we intend to publish a fuller account in an appropriate publication. In developing this model there were a number of general considerations that guided the modeling. First, there was the desire to be able to relate the various elements of the model to specific physiological entities and to give them physically realistic values. This procedure has heuristic value and has a maximum chance of providing physical insight. Furthermore, this procedure should allow modeling of the ears of more than one species, each with its own characteristic structures and values. Such a capacity is desirable because certain kinds of data are available on only some species because of anatomical, practical or ethical considerations. Secondly, the model should be calculable with modest computer resources. Third, the main focus of the model was on the behavior of the ear at high intensities where loss was thought to be a function of mechanical stress (130 dB and up). And lastly, the model had the goal of taking free-field sound pressures as the acoustic input and passing the energy through the outer, middle and inner ears to ultimately provide an estimate of stress within the sensory cells of the inner ear (the organ of Corti). These last goals required that models of the head, outer ear, middle ear, and inner ear all be integrated and allowances made for non-linear elements in them.

Assuming that the foregoing efforts would meet with some success, we hoped to 'explain' certain perplexing data on the ear's susceptibility. Specifically, research in two different laboratories has shown the ear to be most susceptible to impulses with spectral peaks in the mid-range and much less susceptible to impulses with their peak energy at lower frequencies (Dancer, Lenoir, Buck and Vassout, 1983; Price 1983a; 1986b). In experiments with the cat ear and impulses from weapons, howitzer impulses were found to be as hazardous as rifle impulses only when the howitzer impulses were about 10dB more intense (Price, 1983a; 1986b). Furthermore, losses tended to be in the middle range of frequencies for both types of impulse (Price, 1983a; 1986b; Price and Lim, 1984). These findings are crucial to the rating of hazard, because all DRCs would have predicted essentially the opposite result. Furthermore, no traditional measures of sound (energy, A-weighted energy, etc.) produce the same ordering of effect.

In this paper we will describe the models used for each of the elements in this integrated chain, we will compare the model's output with appropriate data, and lastly, we will compare the model's output with various weapons impulses.

THE MODEL

In describing the model, the details presented will be appropriate for the cat ear. This choice is made because the physiological and acoustic values are best known for the cat and also because the hearing loss data to be explained were produced with the cat ear. The ears of other mammals are similar and the principles used in the modeling should transfer with only modest adaptations to fit anatomic details. A schematic diagram of the human ear with the anatomical elements labeled is provided in Fig. 1 as an aid in following the development of the model.

Given the large number of authors who over the years have proposed models for various parts of the auditory system, it is difficult to parcel out the

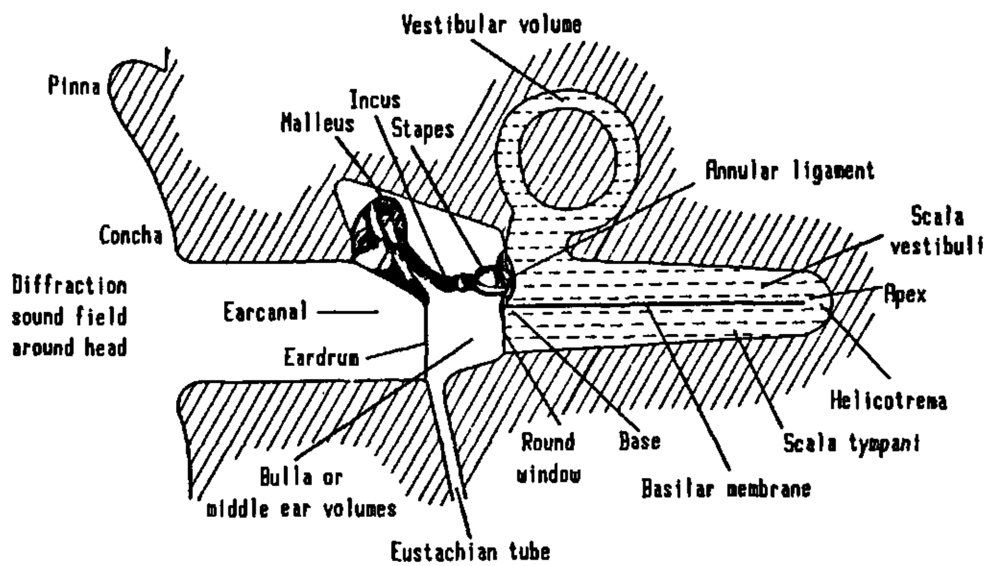

Fig. 1. Schematic diagram of the ear.

intellectual debt accurately. In producing this synthesis the authors gratefully acknowledge the efforts of the many who have preceded them and apologize for any citations that are missed.

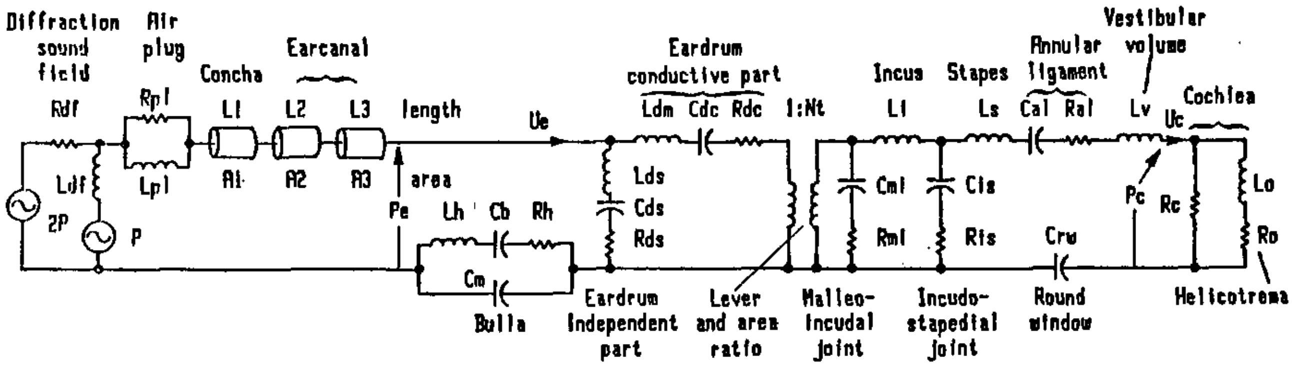

Fig. 2. Schematic diagram of the ear model. P is the free-field sound pressure; Pe is the pressure at the eardrum; Ue is the volume velocity into the eardrum. Pc is the pressure across the BM at the base; Uc is the volume velocity into the cochlea at the oval window. The values of the elements for Bulla closed are: Rdf = rho*c/Shead; Ldf = 0.5*rho/SQRT(PI*Shead); rho = 1.13E-3 gm/cm^3; c = 3.52E4 cm/sec; Shead = PI*Rhead*Rhead; Rhead = 1.65 cm; Rpl = rho*c/S1; Lpl = 0.85*rho/SQRT(PI*S1); L1 = 1.185 cm; L2 = L3/3; L3 = 1.705 cm; S1 = 5.065 cm^2; S2 = 2*S3; S3 = 0.375 cm^2; Lds = 0.0482; Cds = 4.70E-7; Rds = 715; Ldm = 0.0214; Cdc = 3.62E-7; Rdc = 60; Rh = 20; Lh = 0.01348; Cb = 3.806E-7; Ct = 1.875E-7; Nt = 53; Cmi = 7E-12; Rmi = 6.4E5; Li = 1.6; Cis = 8.4E-12; Ris = 1E4; Ls = 3.3; Cal = 1.11E-9; Ral = 2E5; Lv = 22; Ro = 1.2E6; Rc = 2.8E5; Lo = 2250; Crw = 1.0E-8; PI = 3.1415. In case the bulla is opened but the bony septum is not removed, three elements change in the bulla model: Cb(open) = Cb*4.70; Ct(open) = Ct*1.06; and Lh(open) = Lh*1.16. In case the bony septum is removed also, then the bulla impedance becomes negligible and is short-circuited in the model. The units of all masses (inductors) are gm/cm^4; viscous loss elements (resistors) are dyne-sec/cm^5; and compliances (capacitors) are cm^5/dyne. The formulas for the tube sections in the outer ear are:

$$P_o = P_i \cos(kL) + i\, U_i Z_c \sin(kL)$$

$$U_o = U_i \cos(kL) + i\, P_i Z_c^{-1} \sin(kL)$$

where:

P_i = pressure magnitude at tube input end.
U_i = volume velocity magnitude at tube input end.
P_o = pressure magnitude at tube output end.
U_o = volume velocity magnitude at tube output end.
Z_c = rho*c/S , characteristic impedance of tube.
S = cross sectional area of tube.
L = length of tube.
k = 2*PI*f/c , wavenumber of waves in tube.
f = frequency in Hz.

Fig. 2 shows both a schematic diagram of the ear model along with acoustic and electrical quantities which can be measured at each location. At each stage the energy is filtered and transformed as it passes through. Acoustic vibration in the outer ear becomes mechanical vibration in the middle ear, which becomes fluid vibration in the inner ear, which ultimately becomes electro-chemical energy in the sensory cells and nerve fibers.

From The Free-Field to the Eardrum

The head is modeled as a spherical baffle around which the sound wave diffracts and in which the entrance of the earcanal is located. The field surrounding the head and in the concha (the major cavity in the external ear) is approximated by a simple network proposed by Bauer (1967). Next in sequence is Wiener, Pfeiffer and Backus' (1965) two tube model of the earcanal which carries the energy as far as the eardrum. Component values (Fig. 2) in the model were adjusted to fit the data from Wiener et al. (1965) on the pressure transfer from the free-field to the eardrum (Fig. 3a). The agreement between the model and the data is good for frequencies up to 10 kHz. We believe the lack of agreement at higher frequencies is due to measurement difficulties inherent at high frequencies.

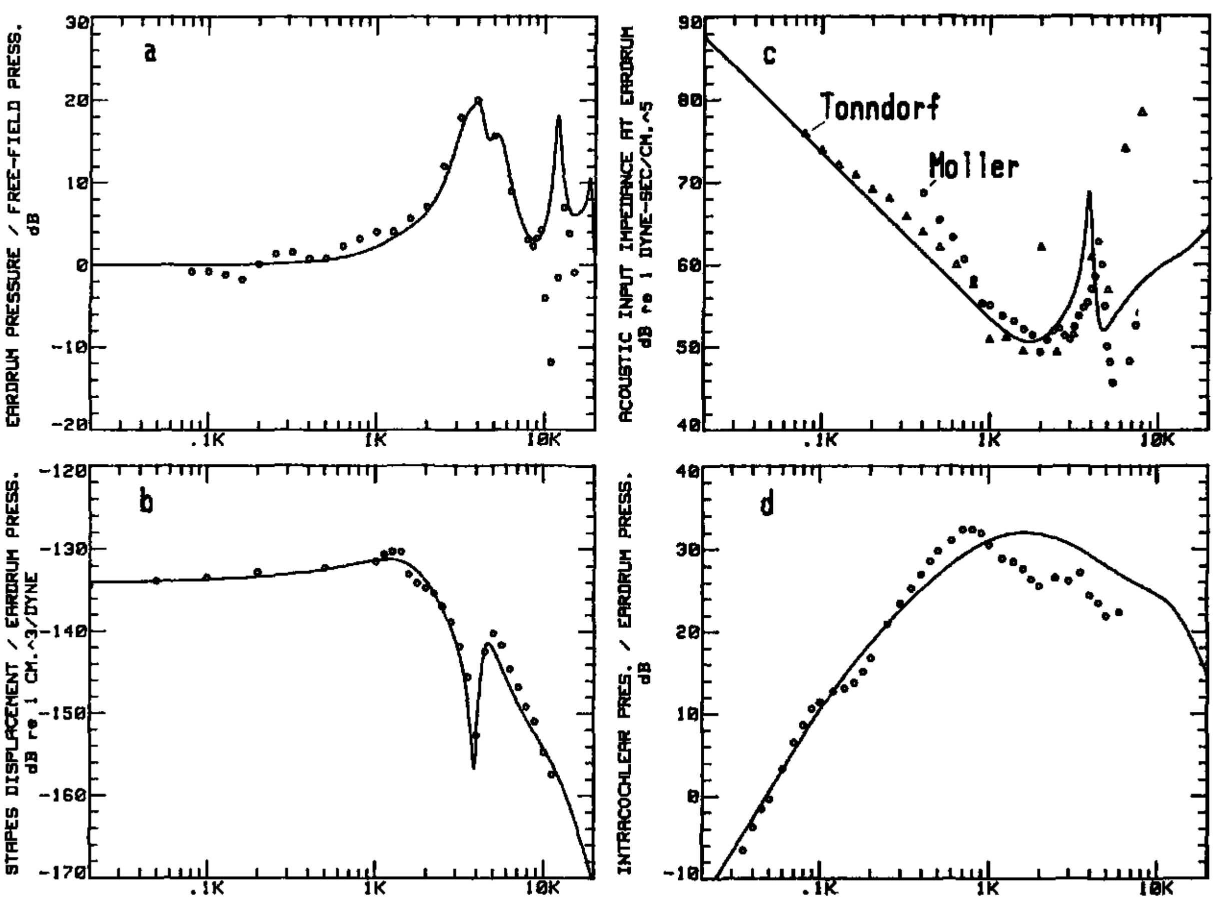

Fig. 3. Comparison of model-calculated values with measurements of four acoustical quantities.

Across the Middle Ear

The essential function of the middle ear is to match the low impedance of air to the high impedance of the cochlea. The model at this point is based on one by Zwislocki (1962) with additional impedance elements given by Lynch (1982). The eardrum is a complex structure which does not move as a piston but actually vibrates in segments. The part of the eardrum that conducts sound to the malleus is represented by a series arm while the part which moves independently is described by a shunt arm. Losses at the ligaments joining the ossicles are also represented by shunt elements. A transformer models the acoustic transformations due to lever action of the ossicles and the ratio of areas of the eardrum and stapes. The connected volumes of the bulla located behind the eardrum contribute to its stiffness and produce a resonance and anti-resonance near 4 kHz described by a parallel-series element. Experiments are often conducted with the middle ear opened. In this case, increasing the value of the bulla elements (Fig. 2) allows the model to account for the middle ear change. The annular ligament which surrounds the stapes at the entrance of the cochlea is described by a visco-elastic series element. The fluid in the vestibular volume is given by a mass element while the round window is modeled as a compliance element. Finally, the fluid load of the cochlea and the low-pass pressure relief of the helicotrema are described by mass and resistance elements.

Model predictions are compared with several acoustical measurements at the eardrum shown in Fig. 3b, 3c, and 3d. The prediction of the ratio of stapes displacement to eardrum pressure (Fig. 3b) is seen to agree very well with the data from Guinan and Peake (1967). They show a uniform response up to 1 kHz and a 12 dB/oct rolloff at higher frequencies. In Fig. 3c we see that the model (solid line) reproduces the input impedance at the eardrum as measured by Moller (1963) and Tonndorf and Khanna (1967). At frequencies above 4.0 kHz there are large discrepancies between measures of the transfer function largely because of the practical problems associated with making the measurements. The wavelength of the sound at these frequencies becomes short enough to be a problem, the eardrum is really an extended surface with a complex shape. Furthermore, at high frequencies it does not vibrate as a single unit but breaks up into complex modal patterns. Finding a single point to make a representative measurement is not really possible. It is therefore not surprising that experimental measurements of acoustic values for the middle ear and external ear might show considerable divergence at specific points (even without considering the problem of individual differences between animals). Fig. 3d shows the ratio of eardrum pressure to intracochlear pressure. The model matches Nedzelnitsky's (1980) measurements reasonably well.

The Cochlea

The cochlea is modeled as a two-chambered, fluid-filled, box with rigid side walls. The partition separating the chambers is rigid except for a tapered basilar membrane (BM) which is narrow at the base and widens exponentially towards the apex. Its mechanical properties also change along its length; the mechanical stiffness per unit area decreases exponentially and the mass and viscous loss per unit area increase exponentially along its length. The mass includes the membrane, the organ of Corti, and the fluid which moves along with it. In contrast to the basilar membrane, the cross-sectional area of the cochlea decreases exponentially from base to apex. At any given location along the BM, we assume that the pressure is constant everywhere within the cross section. This is equivalent to assuming the wavelengths on the BM are long compared to the height of the chambers.

For sinusoidal stimulation at the stapes, a traveling wave appears on the BM. For each frequency, the BM displacement reaches a maximum at a characteristic location, high frequencies having their maxima at the stiffer, basal end of the cochlea and low frequencies having their maxima at the more compliant, apical end of the BM. Given the presence of a traveling wave within the ear, we have solved for the BM motion with a mathematical procedure known as the WKB method (Zweig, Lipes and Pierce, 1976) which is applicable for systems with traveling waves, provided the wavelength is short compared to the dimensions of the system. This limitation of the WKB method is admittedly a problem as the driving frequency gets lower. However the method offers the overriding advantage that closed-form analytic solutions are obtainable.

It was mentioned earlier that the hypothesized hearing-loss mechanism at high intensities was stress within the structures in the inner ear. However, given the limitations of knowledge about the micromechanics of the cochlea, we have accepted BM displacement as a first approximation. The stapes to basilar membrane displacement ratio has been calculated for the 7.4 kHz 'place' on the basilar membrane for tones with SPLs ranging from 55 to 105 dB (Fig. 4). Also plotted for comparison in Fig. 4 are Rhode's (1980) data for the displacement at the equivalent cochlear place in the squirrel monkey cochlea. Two things are apparent. First, the calculated displacement ratios are a reasonable approximation of the measured ratios. Secondly, that at low levels, the response is more peaked than it is for high levels of stimulation. The fit to the data was produced by changing the damping factor used in the calculation (damping values listed on the figure). A more easily interpretable picture of what happens at a particular cochlear location as intensity rises is shown in Fig. 5 which includes essentially the same information as Fig. 4, except that displacements are plotted instead of ratios.

Recent research has suggested that the there is an 'amplifier' in the organ of Corti. If this amplifier worked at low sound intensities by reducing losses in the basilar membrane (smaller damping factor, better tuning) then the sensitivity of the ear would be enhanced when it was needed. Conversely, at high intensities, the effect of the cochlear amplifier would be overwhelmed by the relatively large forces/displacements and the response of the BM would be a function of the essential properties of the membrane/fluid system itself. Therefore in working with noise exposures and this model, we have used the higher damping values appropriate to high levels.

The plot of predicted basilar membrane displacement along its length for pure tone inputs at the stapes, is shown in Fig. 6. The envelopes of traveling wave displacements in Fig. 6 match essential features of what is known about the ear's response. First, the location of the peak varies as the logarithm of the frequency, matching the map produced for the cat cochlea by Liberman (1982) and 'corrected' by the reduced damping noted in the previous paragraph. Second, the amplitude of the peak increases by 3 dB for each octave increase in frequency. This value is close to that observed by von Bekesy (1949). Third, the Q of the peaks increases with frequency.

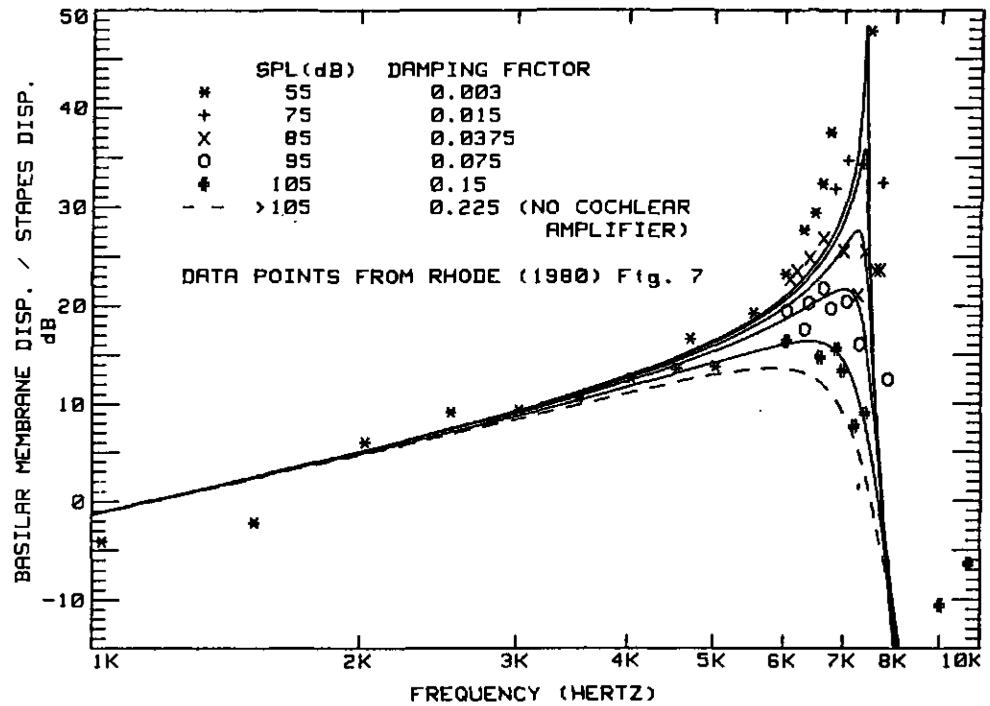

Fig. 4. Comparison of model calculation of basilar membrane to stapes displacement ratios at the 7.4 kHz place for different driving frequencies and intensities. The data points are from Rhode's measurements on the squirrel monkey (1980).

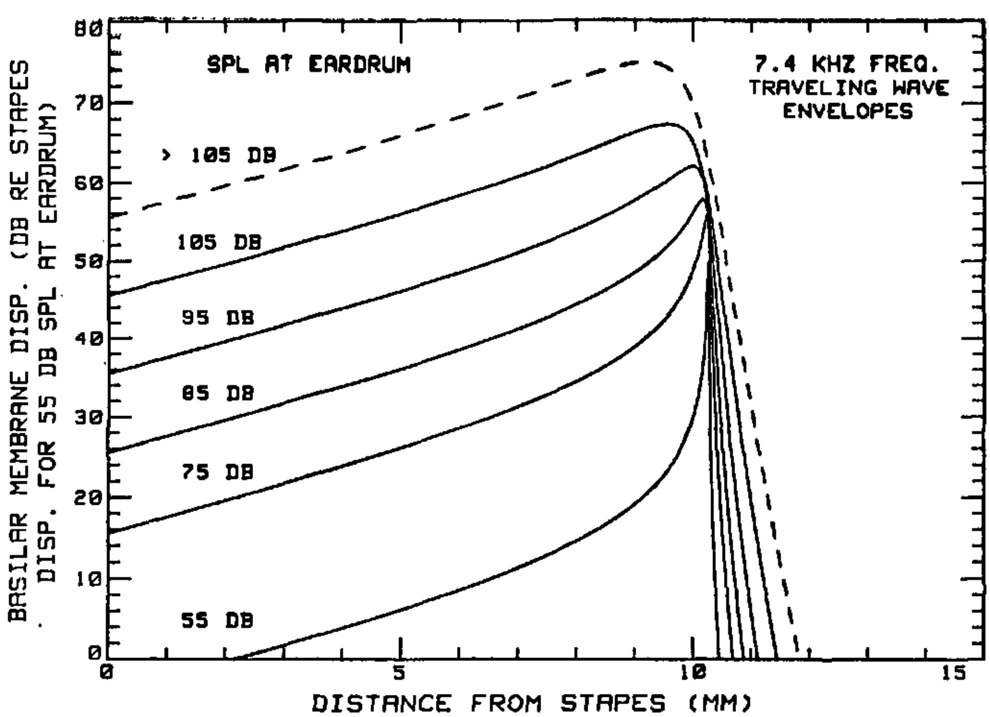

Fig. 5. Model calculation of the basilar membrane displacement envelope at the 7.4 kHz place and sound pressures from 55 dB to over 105 dB.

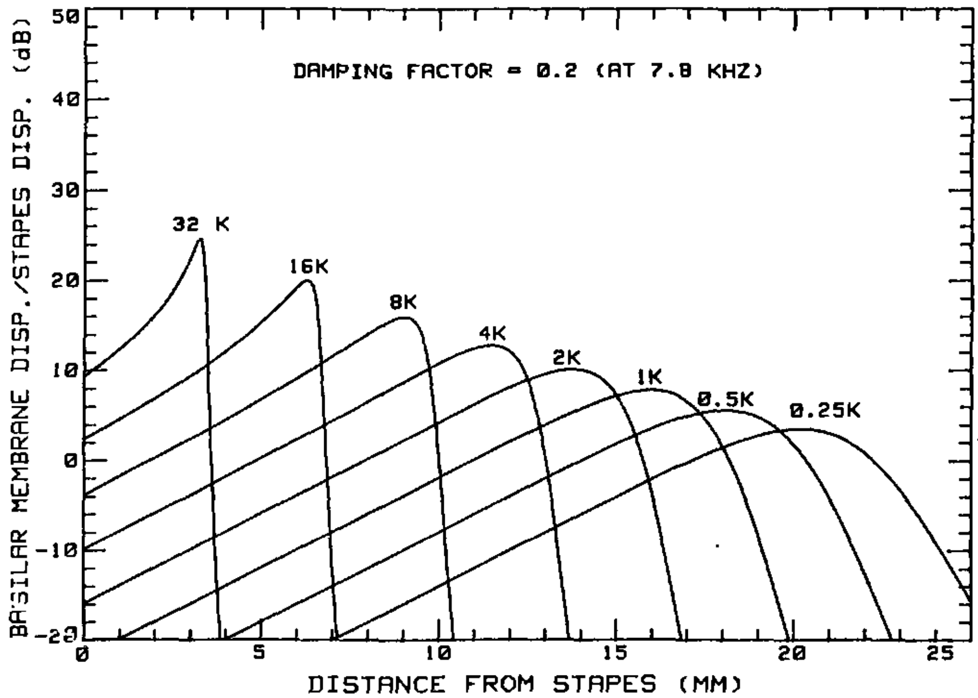

Fig. 6. Calculated envelope of traveling wave displacements for different frequencies.

The Stapes

Thus far, the components of the model have been linear elements. At high levels, however, the annular ligament of the stapes introduces important non-linearities, (Price, 1974). The ligament is rugged and is linear for almost all sound pressures to which the ear is likely to be exposed (Wever and Lawrence, 1954). However, at very high intensities the ligament limits stapes motion, effectively clipping the cochlear input. Guinan and Peake (1967) published data on stapes motion at high sound pressure levels which are reproduced in Fig. 7 along with a non-linear model (solid line) fitting the data. The ligament model predicts that stapes displacements will reach an asymptote at about 42 microns peak to peak displacement. The solid line of the model fits the displacement data very well. The sound pressure required to reach the displacement limit will of course be a function of the frequency of the driving stimulus and has in essence been calculated for pure tone stimulation (Price, 1974). On the other hand, the model makes it possible to do the calculation for spectrally complex stimuli in which we have a practical interest. Wever also showed (pp. 148-151) the eardrum moves linearly for displacements of more than 100 microns; hence allowing for the middle ear lever ratio of about 2.5, the eardrum would not produce any additional distortion.

In applying the clipping calculation to the model, we recognize that the addition of a non-linear element must to some extent be reflected in the responses of other middle ear structures. The present calculational scheme makes no specific allowance for such effects. This may not be too serious a limitation, given that the middle ear is critically damped and does not show appreciable energy storage and ringing, at least at the levels for which

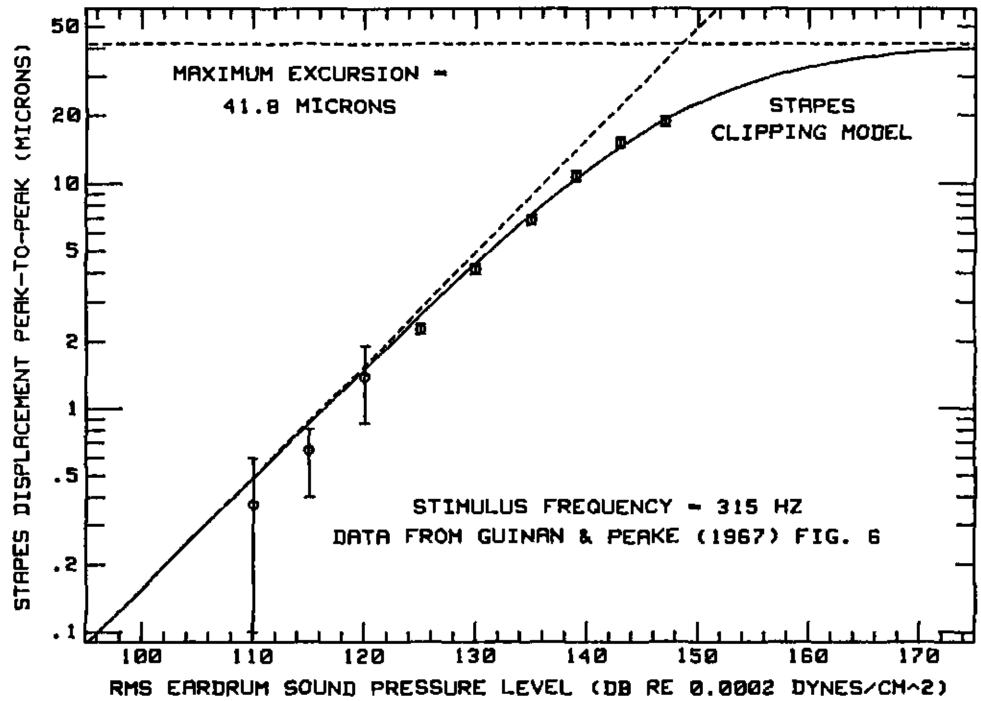

Fig. 7. Comparison of clipping-model output for stapes displacement with empirically determined values from Guinan and Peake (1967).

measurements have been made. The data from Guinan and Peak (1967) in Fig. 7 are consistent with this assumption. Furthermore, in modeling the otosclerotic ear (a condition in which the stapes is immobilized in the oval window), Zwislocki (1962) pointed out that the effect of stapes immobilization on the impedance at the eardrum is negligible, due to the high impedance of the incudo-stapedial joint and the low coupling impedance of the eardrum at high frequencies.

In calculating with the present model, the solution for BM motion is found in three steps. First, the linear solution is found for stapes motion, ignoring annular ligament non-linearities. Next, the linear solution is then non-linearized according to the empirically determined time-domain clipping transfer function of Fig. 7. And third, the modified motion is used as input to the cochlear portion of the model which in turn predicts BM displacements at specific points.

Given stapes displacement as an intermediate point, the transfer function from stapes displacement to BM displacement at a distance x from the stapes for a driving frequency f is D(x,f). The sign convention is that an initial displacement (positive) of the stapes into the scala vestibuli causes a downward (negative) displacement of the BM (from scala vestibuli toward scala tympani). The calculational formula is:

$$D(x,f) = \frac{A_s M}{6.68\ b_o d g}\, e^{-0.09x}\, \frac{Z_x}{S_x} \left(\frac{Z_x S_o}{S_x Z_o}\right)^{1/2} \left(\frac{Z_o + S_o}{Z_x + S_x}\right)^{M}$$

where:

A_s = 1.26 mm^2 , area of stapes footplate.
b_o = 0.08 mm , width of BM at stapes end.
d = 4.68 mm , tuning factor of BM.
$delta_o$ = 0.024, damping factor of BM at stapes end.
x = distance from stapes in mm along BM.
N = 3.16 , number of traveling waves on BM at one time.
f = frequency of stapes oscillation in kHz.
f_o = 66.1 kHz , local resonance frequency of BM at stapes end.

f_x = $f_o e^{-x/d}$, local resonance frequency of BM at location x.

$delta_x$ = $delta_o e^{x/d}$, damping factor of BM at location x.

$i = (-1)^{1/2}$
$g = (1 - i\ delta_x)^{1/2}$
$M = 4N/g$
$Z_o = g\ f/f_o$
$Z_x = g\ f/f_x$

$S_o = (Z_o Z_o - 1)^{1/2}$

$S_x = (Z_x Z_x - 1)^{1/2}$

The empirical formula for converting the linear solution for stapes motion into an approximate non-linear solution is:

$$x_{nl} = \frac{x_{lin}}{1 + |x_{lin}/x_{max}|}$$

where:

x_{lin} = linear stapes displacement in microns; calculated from the linear ear-model with no limitation by the annular ligament.
x_{lin} = P_e*A_{lf}, in the case of the low frequency tone used by Guinan and Peake.
A_{lf} = 0.142E-6 cm^3/dyne, low-frequency closed-bulla transfer ratio of the middle ear listed in Table II of Guinan and Peake.
P_e = magnitude of sound pressure at eardrum.
x_{max} = 20.9 microns , maximum stapes displacement extrapolated from pure tone data of Guinan and Peake assuming symmetrical peak clipping.
x_{nl} = non-linear stapes displacement in microns.

APPLICATION OF THE MODEL

Work in applying this model has just begun; therefore many details of its application remain to be explored. However, the early results are promising. The model does provide plausible explanations for the two major 'anomalies' in the hearing loss data, namely, the fact that for equal peak pressure levels (PPLs) the rifle is more hazardous than the howitzer and the fact that loss occurs in the mid-range of frequencies, even when the spectral peak is elsewhere.

The prediction of loss is the mid-range was made some years ago (Price, 1977; 1981) based primarily on the tuning of the external and middle ears and the relatively flat spectra of commonly encountered noises. Because the present model embodies the same transfer functions, it is also consistent with the earlier contentions. Later studies with noise exposures have shown that exposures to impulses from both the rifle and howitzer produce histologically and audiometrically measured damage to the middle of the cochlea (Price, 1983a; 1986b; Price and Lim, 1984)

The more puzzling problem was the prediction of the relative ranking of the hazard from large and small caliber weapons. The explanation that has resulted from work with the model is presented in the next two figures. Fig. 8a shows the stages in the ear's response to a rifle at 145 dB PPL. This impulse and the others used here are those actually recorded during impulse noise exposures and include all the features normally found in this type of data, such as the ground reflections. The top panel shows the free-field pressure, typical of a rifle. The middle panel shows the corresponding stapes displacement. The largest displacement was to the positive phase of the impulse and was about 7 microns, the ligament model clipping only slightly. The lowest panel shows the predicted BM displacement at the 3 kHz place in the cochlea, the area where the largest effects have been noted. Note that the largest displacement of the BM was about 12 microns upward and is in response to the <u>rarefaction</u> phase of the waveform. The parallel set of data is shown in Fig. 8b for the rifle impulse at 155 dB PPL. The stapes displacement showed some clipping (it actually rose less than 6 dB to 13 microns for a 10 dB increase in PPL). On the BM, the largest displacement was still upward and was more than 25 microns.

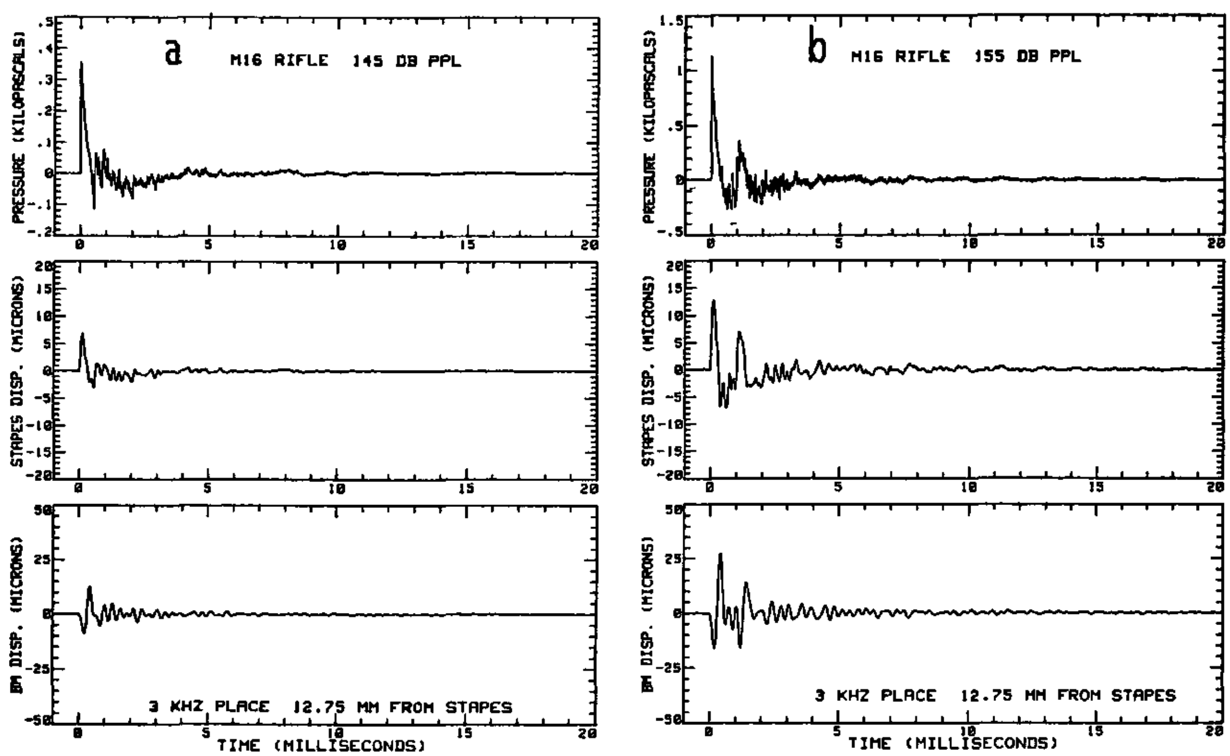

Fig. 8. Displacements within the ear calculated for a rifle impulse at 145 dB and 155 dB PPL.

For the rifle, the largest displacements are in the upward direction. Therefore, we might suspect that scala vestibuli-ward movements do the most damage. We in fact suspect that hazard may well be a function of the direction of movement. Such a contention makes sense physiologically. In essence, downward movement produces essentially compressive stresses whereas upward movement produces primarily tensive deformations. Tissue tends to fail in tension; therefore it is reasonable to suspect that upward BM motions might be the most damaging.

The howitzer impulse in Fig. 9a is 154 dB PPL and produces maximum stapes displacements of about 13 microns, about the same as the rifle at the same PPL. On the BM, the displacements in the downward direction are about the same as for the rifle at the same PPL. However, the upward displacements are <u>smaller</u> than those for the rifle by more than a factor of two. In Fig. 9b, the calculated response to the howitzer impulse at 166 dB can be seen to produce stapes displacements of almost 20 microns inward and a little less than 15 outward. If no clipping had occurred, the peak inward displacement would have been over 160 microns! The clipping introduced by the annular ligament of the stapes obviously has a major effect on what gets transmitted to the inner ear at high PPLs. The calculated BM motions make an interesting pattern. Perhaps the most important point is that the maximum upward displacements are on the order of 25 or 30 microns, about the same as for the rifle when it was 11 dB less intense. Furthermore, the largest displacements occur not in response to the fundamental waveform; but to the 'hash' riding on it. It is these smaller oscillations that produce large stapes movements, and if their timing is correct, the BM at the 3 kHz place responds. It is interesting to note that an intense impulse with a long A-duration would push

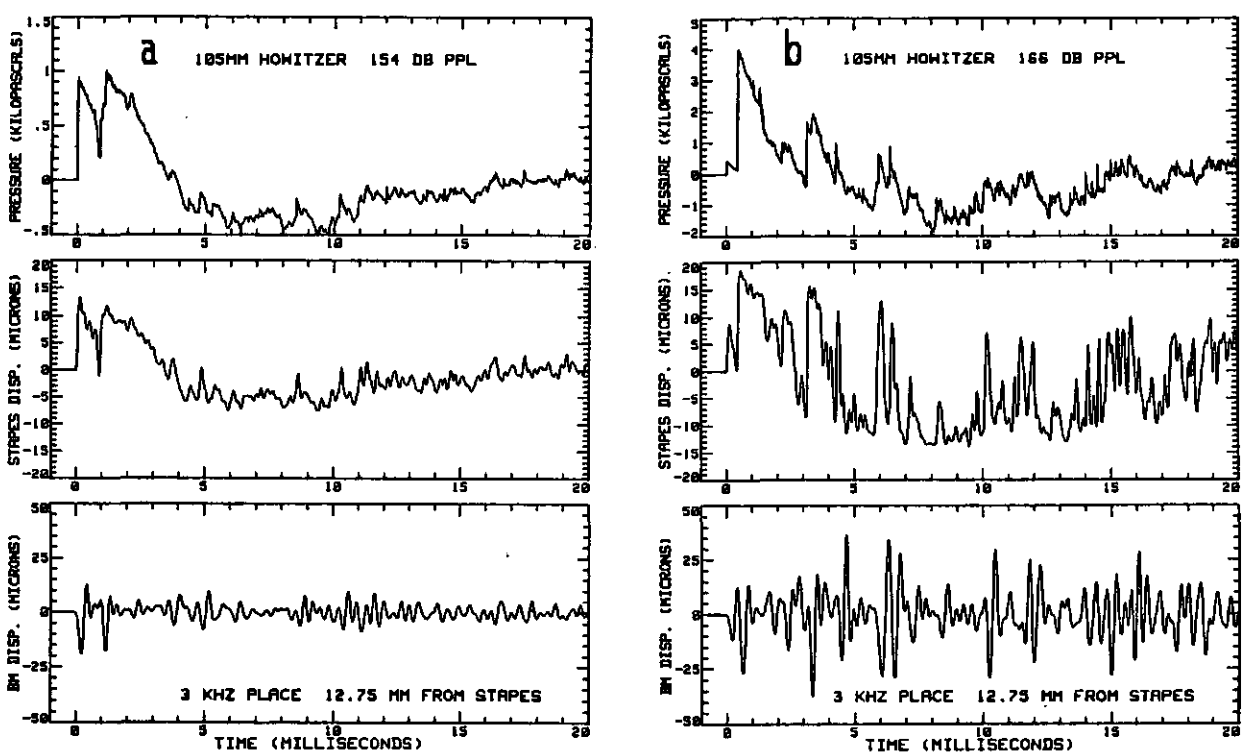

Fig. 9. Displacements within the ear calculated for a howitzer impulse at 154 dB and 166 dB PPL.

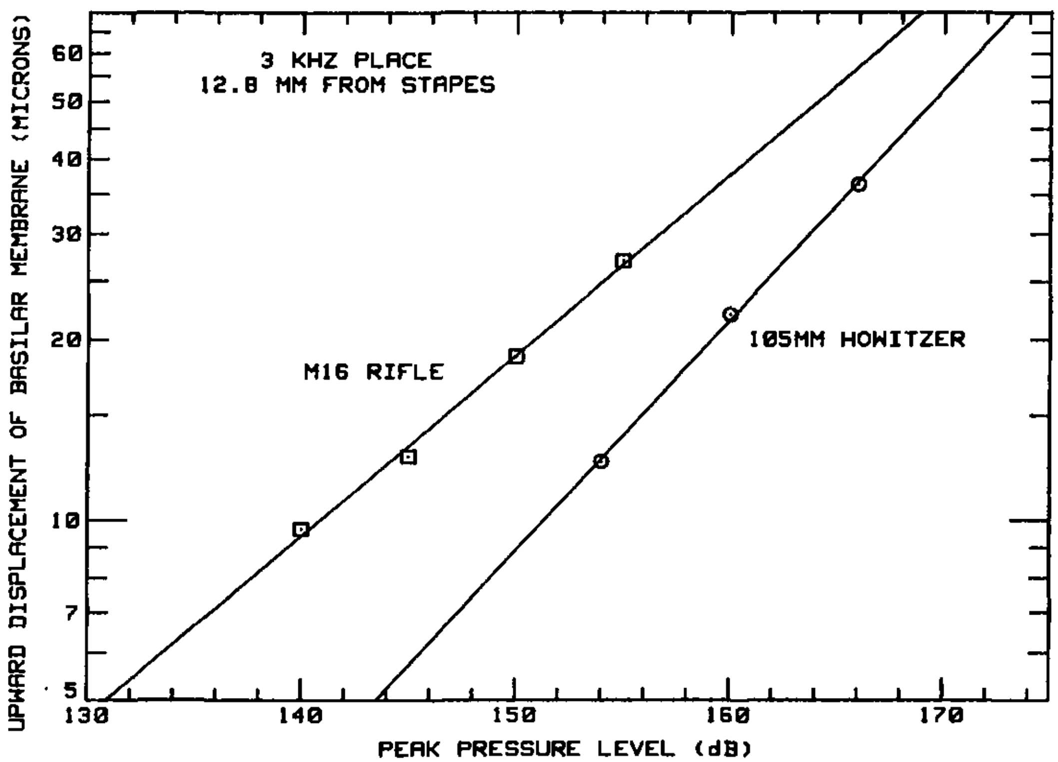

Fig. 10. Upward BM displacement as a function of PPL for howitzer and rifle impulses.

the stapes in and hold it there, effectively blocking transmission into the inner ear during that period.

Fig. 10 summarizes what we believe to be the major point of the last two figures. The upward displacement of the BM as a function of PPL grows as PPL rises for both impulses, a little faster for the howitzer than for the rifle. However, note that for equal displacements, the PPL of the howitzer must be higher than that for the rifle by about 10 dB. This is the ranking of hazard that the hearing loss measurements have produced.

These results are encouraging in that they do seem to predict what the hearing loss data are showing and harmonize all the data. And if we speculate freely, the model could be used as a basis for a convenient 'hazard meter' for any type of intense sound. Furthermore, the model has heuristic value in that it can be used to calculate ways of changing the wave form to produce a safer impulse. What is needed now is confirmation of the model's predictions in real ears.

We intend next to extend the model to include different impulses on which we have hearing loss data, to generate predictions of hazard for different species to include man, and to generate testable hypotheses as a means of verifying the essential elements of the model. What is reported here is a preliminary effort; but if it is supported by subsequent research, it is fair to say that the assessment of impulse noise hazard will undergo dramatic changes.

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