Bio- REHABILITATION
ENGINEERING MODULES

ClD-U-08-CU

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GATEWAY COALITION

THE COOPER UNION FOR THE ADVANCEMENT OF SCIENCE AND ART

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PHYSIOLOGY OF HEARING AND PSYCHOACOUSTICS

Human Hearing

The mechanism of the human ear has been a source of much wonder for physiologists, who are progressing well beyond the fragmentary knowledge of the past by continually uncovering layers of new marvels of how the human ear really functions. Our hearing mechanism is a complex system that consists of many subsystems. The mechanism of the brain's processing of auditory stimuli is probably beginning to be understood, with the relatively recent discovery that a part of the hearing system is metabolic in nature. Much of the pioneering work was performed by Georg von Bekesy who received the Nobel Prize for his investigations, particularly those entailing the frequency selectivity of the inner ear. His work indicated that the frequency selectivity ranked far poorer than the ear actually exhibits, but William Rhode found much greater selectivity in his work with live animals (von Bekesy worked with dead animals, which most likely accounted for the difference). It is now realized that frequency selectivity of the inner ear fades within minutes after the metabolism ceases.

A young, healthy human is capable of hearing sounds over the frequency range of 20 Hz to 20 kHz, with a frequency resolution as small as 0.2%, Thus, we can discern the difference between atone of 1000 Hz and one of 1002 Hz. With normal hearing, a sound at 1 kHz that displaces the eardrum less than 1 Angstrom can be detected, in fact, less than the diameter of a hydrogen atom!. The intensity range of the ear spans extremes from threshold (at which softest sounds can be detected) to the roar of a fighter jet taking off, thus covering an intensity range of approximately 100,000,000 to 1. The ear acts as a microphone in the process of collecting acoustic signals and relaying them through the nervous system into the brain. The ear (cf. Figure 10-1) subdivides into three principal areas: the *outer, middle* and *inner* ear. The outer ear consists of a *pinna* which serves as a soundcollecting horn and the auditory canal that leads to the inner ear. The collected sound enters the ear through the opening (the *meatus)* into auditory canal which forms a tube approximately 3/4 cm in diameter and 2.5 cm length. The canal terminates at the *tympanic membrane* (eardrum). Under the impetus of sound the eardrum vibrates, causing three bones linked in a ossicular chain---namely the *malleus* (or hammer), the *incus* (anvil) and the *stapes* (stirrup)---to oscillate sympathetically. At the lowest resonance (3 kHz) of the auditory canal, the sound pressure level at the eardrum is about 10 dB greater than it is at the entry into the canal. Because a resonance curve tends to be broad, human hearing tends to be more sensitive to sound in the range of approximately 2 to 6 kHz, as the consequence of the resonance being centered at 3 kHz. The diffraction of sound waves inside the head has the effect of causing the sound pressure level at the eardrum to exceed the free-field sound pressure level by as much as 20 dB for some specific frequencies.

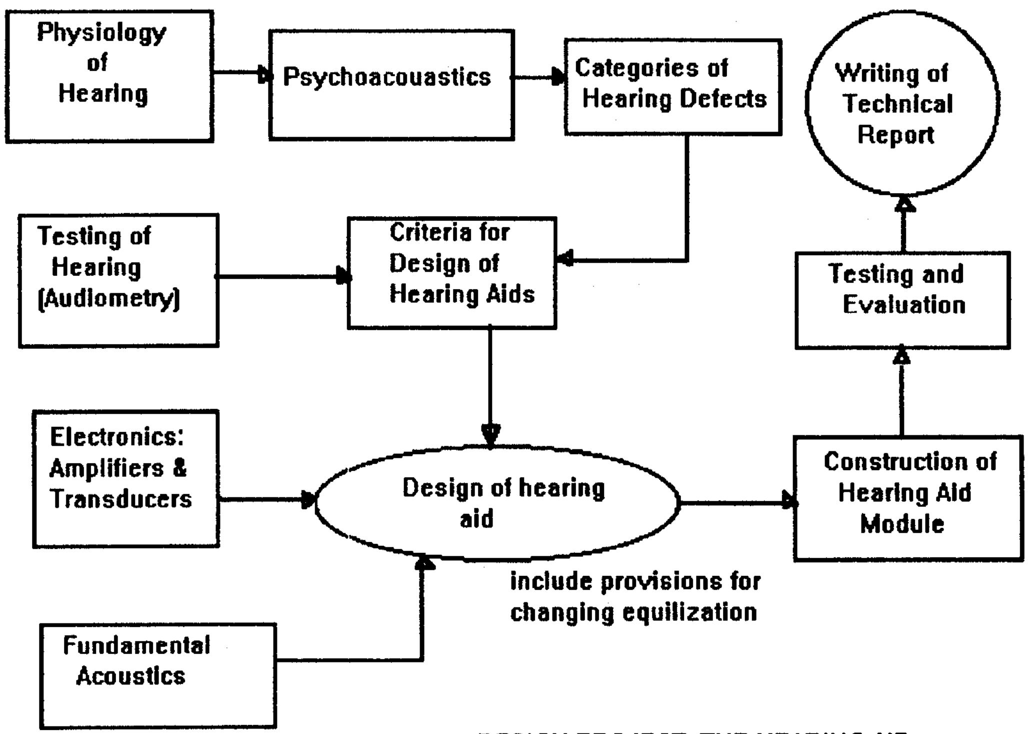
The eardrum itself is a thin, semitransparent diaphragm the completely seals off the canal, marking the inner boundary of the outer ear and the outer boundary of the middle ear. This membrane is quite flexible at its center and attached at its perimeter at the terminus of the auditory canal, thus constituting the entrance to the middle ear. The middle ear, lined with a mucous membrane, constitutes an air-filled cavity of about 2 cm3 in volume, which contains the three ossicles (bones), namely the malleus, incus and the stapes forming a bony bridge from the external ear to the middle ear. These bones are supported by muscles and ligaments. The malleus is attached to the eardrum; the incus connects the malleus and the stapes. The last bone in the chain, the stapes, covers the *oval window.* The *Eustachian tube,* which is normally closed, opens in the process of swallowing or yawning to equalize the air pressure on each side of the eardrum; this is a tube approximately 37 mm in length that connect the middle-ear cavity with the pharynx at the rear of the nasal cavity.

Just below the oval window lies another connection between the middle and inner ears, the membrane-covered *round window.* Between the oval and round windows is a rounded osseous projection, formed by the basal turn of the cochlea, called the *promontory. A* canal encasing the facial nerve is situated just above the oval window.

The structures to the right of the oval and round windows in Figure 10-1 are collectively called the *inner ear* (also called *labyrinth),* which is comprised of a number of canals hollowed out of the petrous portion of the temporal bone. These interconnecting canals contain fluids, membranes, sensory cells, and nerve elements. Three principal parts exist in the inner ear: the *vestibule* (an entrance chamber), the *semicircular canals,* and the *cochlea.* The vestibule connects with the middle ear through the oval and the round windows. Both of these windows are effectively sealed, by the action of the stapes and its support on the oval window and the presence of a thin membrane in the round window, thus preventing the loss of the liquid filling the inner ear. The semicircular canals play no role in the process of hearing but they do provide us with a sense of balance. The cochlea, shown in enlarged detail in Figure 10-2, is the sensory system that converts the vibratory energy of sound into electrical signals to the brain for the detection and interpretation of that sound. The cochlea can be described as a 3.5cm long tube of roughly circular cross section, wound about 2'/2 times in a snaillike coil. This tube's cross sectional area decreases in a **from its base to its apex. Its total volume is about 5 $(10)^{-2}$ cm^3.**

The coils of the cochlea surrounds an area called the modiolus; and the membranous labyrinth of the cochlear sector of the inner ear divides into three ducts or galleries (scalae). The cochlear duct (ductus cochlearis) runs the length of the spiraling cochlea, and because it occupies the central portion of the chochlea's interior, it has been termed the scala medi, whose walls effectively partition the cochlea into two longitudinal channels, the scala vestibuli (or upper gallery) and the scald tympani (lower gallery). , The only communication between the two galleries occurs through the helicotrema, a small opening at the apex of the cochlea. The other ends of the upper and lower galleries terminate in the oval and round windows, respectively.

Figure 10-3 shows an enlarged view of the cochlear duct. This duct is bounded by Reissner's membrane, the basilar membrane, and the stria vascularis.



DESIGN PROJECT: THE HEARING AID

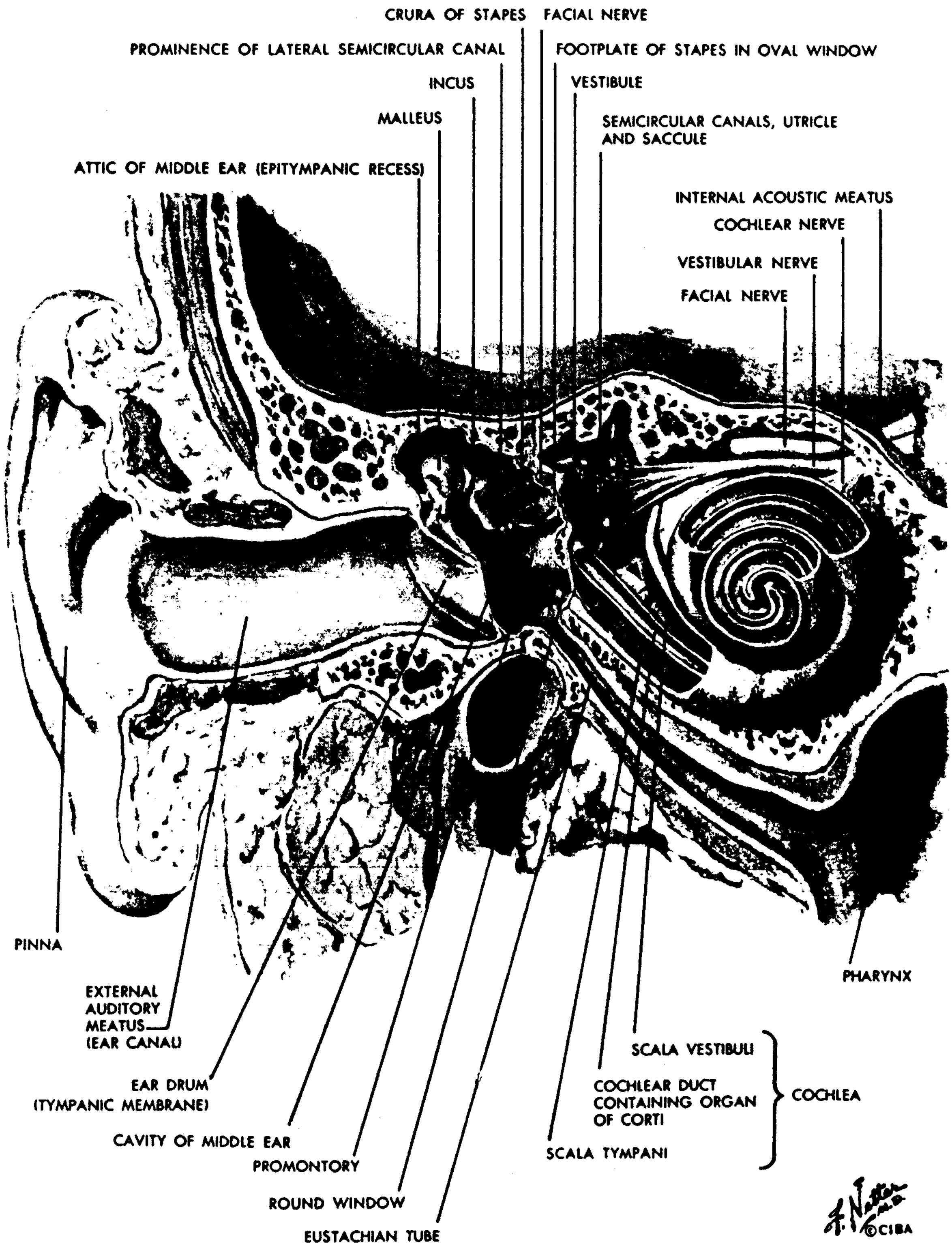


Figure 10-1 Coronal section of the right ear. (From Frank H. Netter, *Clinical Symposia*, © 1970, CIBA Pharmaceutical Co., Division of CIBA-GEIGY Corporation. Reproduced with permission of the publisher.)

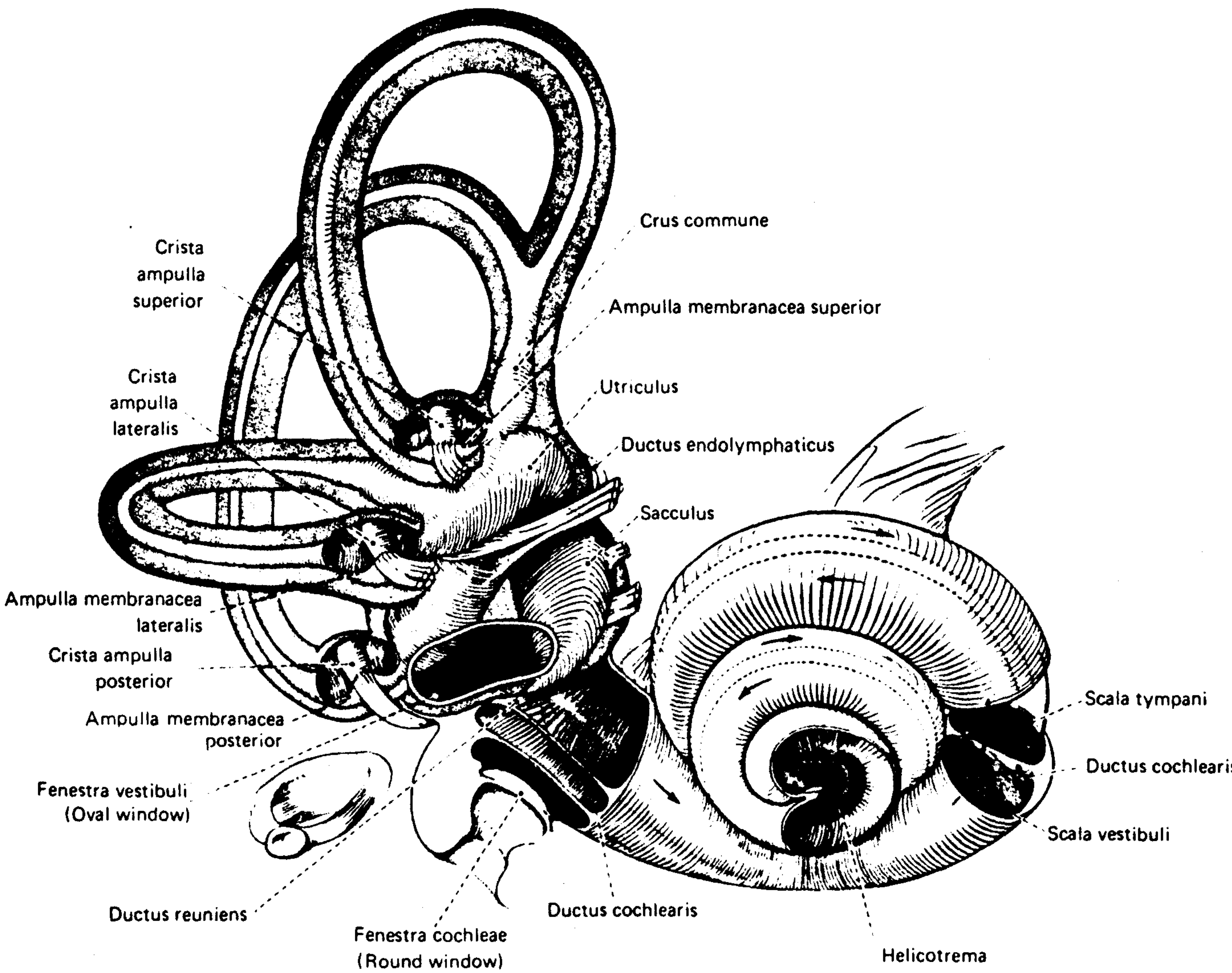


Figure 10-2 The membranous semicircular canals showing the cristae within the ampullae. (From Anonymous, "The Internal Ear," *What's New*, 1957, Abbot Laboratories. Reproduced with permission of the publisher.)

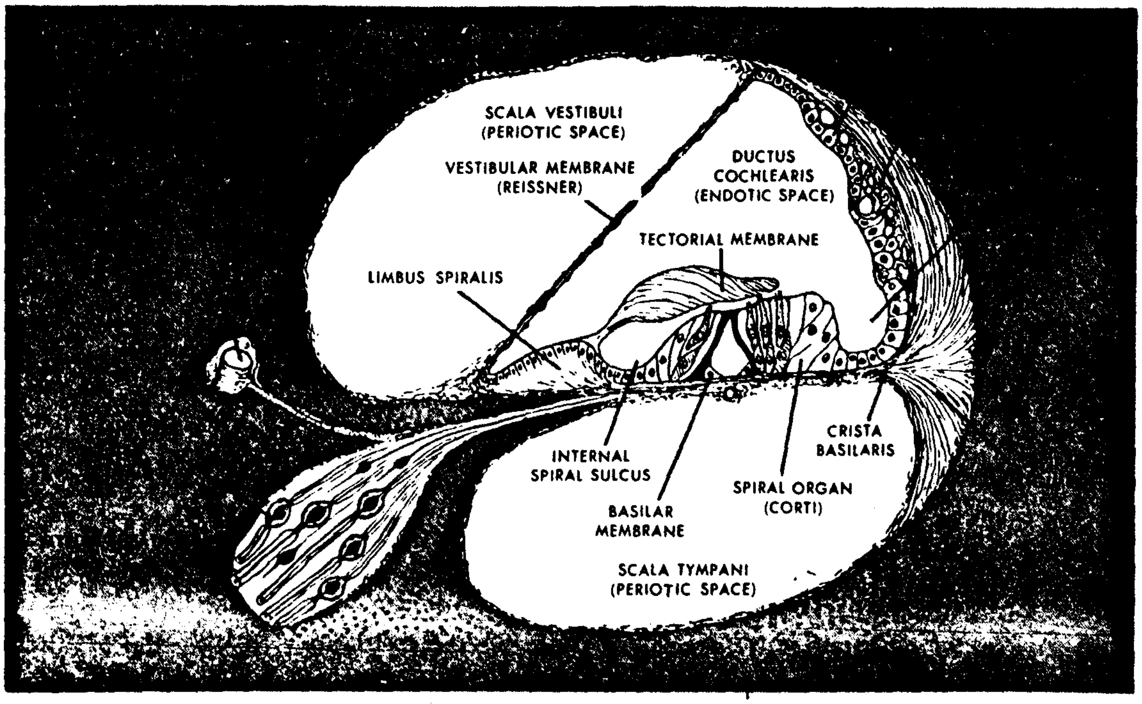


Figure 10-3 Cross section of the cochlear canal. (From A. T. Rasmussen, *Outlines of Neuro-Anatomy*, 3rd ed., 1943, Wm. C. Brown Co.

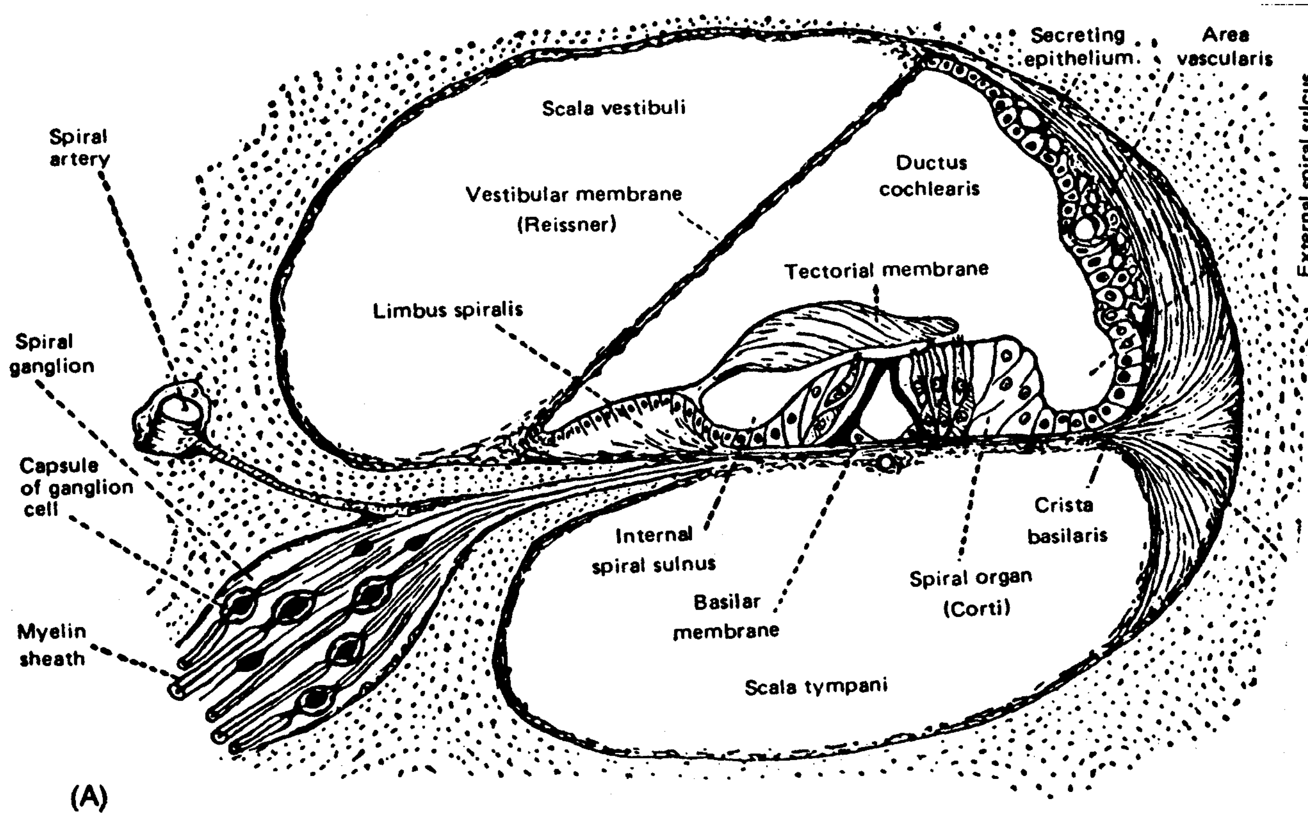
The basilar membrane extends from the bony *spiral lamina,* a ledge extending from the central core of the cochlea, to the *spiral ligament.* The length of the basilar membrane is about 32 mm long, from the base to the apex of the cochlea; the width varies from about 0.05 mm at the base to about 0.5 mm at the apex; and the membrane becomes thinner gradually at it nears the apex. Positioned on the basilar membrane is the *organ of Corti,* This organ, shown in detail in Figure 10-4, consists of some structural cells (e.g. *Dieter's cells* and *Hensen's cells),* the rods and tunnels of Corti, and two types of hair cells on top of which lies the *tectorial membrane.* The tunnel of Corti, isolated from the endolymph, contains the fluid *cortilymph.*

The hair cells constitute the sensory cells for hearing. The *inner hair cells* are arranged in a single row on the modiolar side of the tunnel of Corti, and the *outer hair cells* exist in three parallel rows on the strial side of the tunnel of Corti. The inner hair cells are round and squat; their upper surfaces contain about 50 to 70 hairs called stereocilia. The outer hair cells, which look more reed-like than the inner hairs, contain about 40 to 150 stereocilia arranged in a W-shaped pattern. There are 3000 to 3500 ciliated cells in the single row of inner hair cells and a total of 9000 to 12,000 ciliated cells in the three rows of the outer hair cells.. The hair cells are connect to some 24,000 transverse nerve fiber in a complex network leading into the central core of the cochlea. The nuclei of these nerve fibers form the *spiral ganglion,* which unite to form the *cochlear branch of the VIIIth nerve.* The cochlear branch joins with the vestibular branch to form the *JIM& cranial nerve,* also called the *auditory* or *vestibulocochlear nerve.* The VIIIth cranial nerve along with the VIIth (facial) cranial nerve proceeds in a helical fashion through the *internal auditory meatus* to nuclei in the brain stem. From the brain stem the auditory pathway extends through various nuclei to the cerebral cortex in the temporal lobes of the brain.

The VIIIth cranial nerve is primarily as sensory nerve ---i.e. it conveys sensory information from the cochlea and the vestibular system to

The Mechanism of Hearing

Sound waves are directed by the pinna into the auditory canal. The longitudinal changes in air pressure of the sound wave propagate to the ear drum, causing it to vibrate. Because the handle of the malleus is imbedded in the ear drum, the ossicular chain is set into vibration. These tiny bones vibrate as a unit, elevating the energy from the eardrum to the oval window by a factor of 1.31 to 1. Sound energy is further enhanced by the difference in area between the eardrum and the stapes footplate by a factor of approximately 14. Multiplying this effective areal difference of 14 by the lever action of the ossicular chain (1.3) yields an energy increase of 18.3 to 1, which translates into an amplification factor of 25.25 dB on the SPL scale. The middle ear acts as a transformer, by changing the energy collected by the eardrum into greater force and less excursion, thereby matching the impedance of the air to the impedance of the inner ear's fluid.



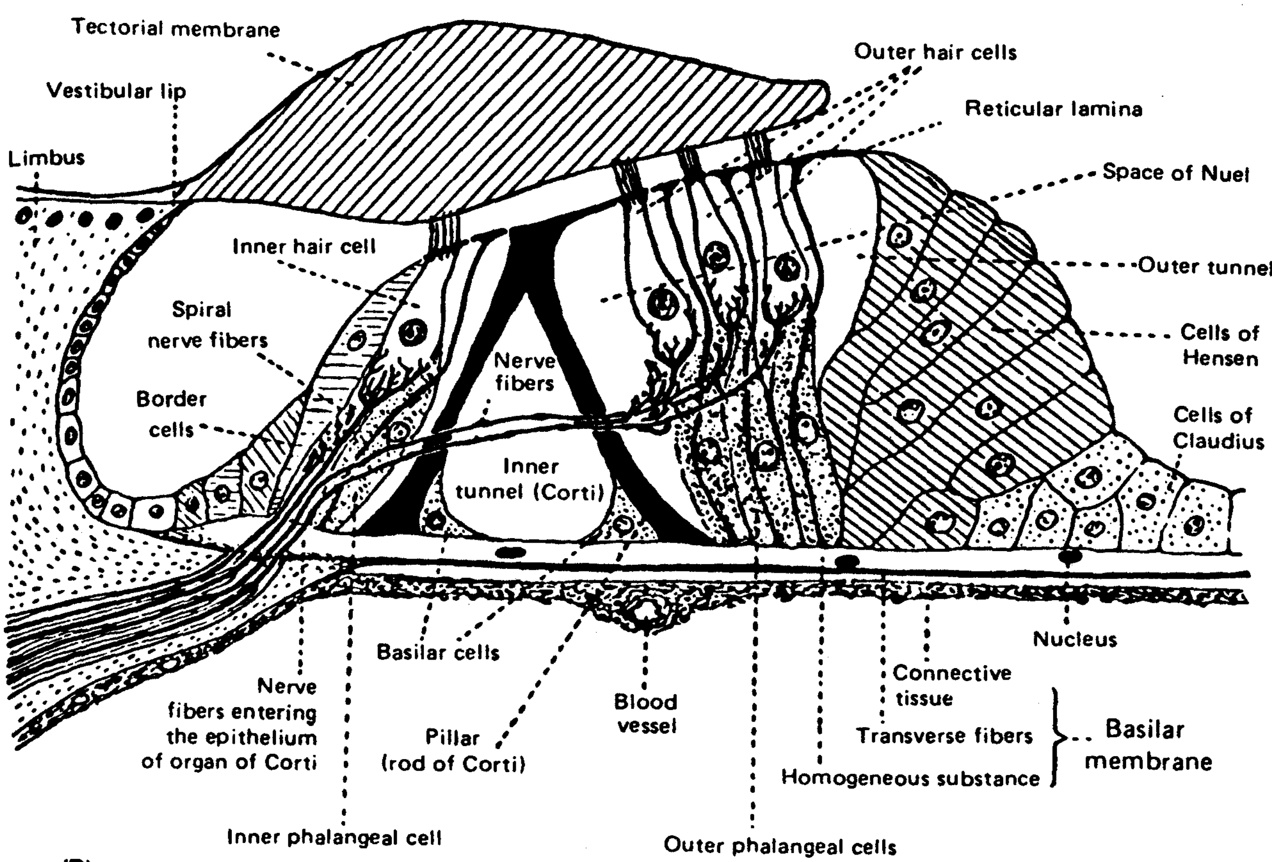


Figure 10-4 A cross-section of the organ of Corti. (A) Low magnification. (B) Higher magnification. (After Rasmussen [1].)

Because the fluid of the inner ear is virtually incompressible, provision for relief of the pressure produced by the movement of the footplate of the tapes is provided by the interaction of oval window and the round window, with the fluid motion from the oval to the round window being transmitted through the cochlear duct. When the footplate of the stapes pushes into the perilymph of the scala vestibuli, the vestibular membrane (or membrane of Reissner) bulges into the cochlear duct, causing movement of the endolymph within the cochlear duct and displacement of the basilar membrane. Von Bekesy's experiments with cochlear models led him to formulate a theory that the displacement of the basilar membrane is in the form of a traveling wave that proceeds from the base to the apex of the cochlea. The maximum amplitude of the wave occurs at a point along the basilar membrane corresponding to the frequency of the stimulus, i.e. each point of the basilar membrane corresponds to a specific value of the simulating frequency.

The cilia of the hair cells are embedded in the gelatinous tectorial membrane, so that when the basilar membrane is displaced, it generates a 'shearing' force on the cilia. The sidewise motion of the cilia creates an alternating electrical current, also referred to as the *cochlear microphonic (CM)*, *cochlear potential* or the *Wever* Bray *effects.* The deflection of the hair cells triggers responses in the neurons connected to the hair cells. Impulses are borne along the nerve fibers to the main trunk of the cochlear portion of the VIIIth nerve and onward to the brain.

1 So named after the two investigators who discovered in 1930 that speech delivered to a cat's ear could be understood when the CM signal was picked up from the cochlear nerve and amplified.

It is thus how the cerebral cortex eventually "hears" the vibration that strike the eardrum.

Research over the past decade indicate that the cochlea does not act passively. Active processes occur that indicate that energy is being added to the cochlea through mechanisms that are not yet fully understood. More energy is contained in the cochlea than that from the sound going into it. One phenomenon that has been identified is that of *otoacoustic emission,* which is a sound in the external ear canal believed to have originated from vibrations within the cochlea and propagated back through the middle ear. Otoacoustic emissions can be measured by placing a miniature microphone in the ear canal. A spontaneous otoacoustic emission is identified as a constant low-level sound that occurs spontaneously in half of normal ears. When a high-level click is introduced to the ear, an *evoked otoacoustic emission* occurs some 5 msec later as a low-level sound. Also, when two different tones are presented at high levels to a normal ear, the otoacoustic emission occurs in the form of new tones generated at frequencies other than the two original frequencies. These new tones are termed *distortion* products.

When the sound striking the eardrum is sufficiently loud, the middle ear muscles contract reflexively. This *acoustic reflex* occurs as a contraction in the stapedius muscle, which results in a pull against the ossicular chain and a reduction in the energy transmitted through the oval window into the perilymph in the vestibule. The largest amount of reduction in sound due to acoustic reflex--from

20 to 30 dB---occurs for low frequencies. Above 2000 Hz the acoustic reflex is fairly negligible.

Hearing Loss

Nearly a quarter of the population between the 15 and 75 years of age surer hearing impairment. This situation renders impaired hearing, which is often caused by infectious diseases or overexposure to loud noise or simply the process of aging, as common as poor vision. When hearing loss occurs in early childhood, its consequences become more obvious than when it occurs in adulthood. A child's progress in learning and developing social relationships may be hindered and the child may even be deemed 'not too bright' if professional help and guidance are not forthcoming. The primary problem of hearing loss, regardless of the age of the affected individual, is a diminished ability to understand speech.

Even milder forms of hearing loss early in life can generate great difficulty, particularly for children who developed within normal limits but are not doing well in school, due to their being inattentive. Such moderate hearing losses are not uncommon and may even be on the increase due to heightened exposure to 'rock' music. When a mild hearing loss is corrected, the child often becomes "like a different person". Fortunately, many of the hearing impaired can be helped through the use of hearing aids.

The diminishing capacity affect adults in a more underhand manner. Most people with age-induced or noise-induced hearing impairment first lose hearing acuity at high frequencies, making it difficult for them to distinguish consonants, especially s versus f, and t versus z. Such persons must strain harder to understand conversations; and going to the movies, listening to lectures, conversing with friends and other pleasures become stressful chores. This can result in an individual's becoming withdrawn from his friends and relatives. Some of these patients can be helped through counseling and rehabilitation, but no cure exists for most cases of sensorineural deafness.

Hearing loss falls into two principal categories: *conductive* and *sensorineural.* Conductive hearing loss occurs from any condition that impedes the transmission of sound through the external or the middle ear. Sound waves are not transmitted effectively to the inner ear because of some blockage in the auditory canal, interference in the eardrum, the ossicular chain, the middle ear cavity, the oval window, the round window, or the Eustachian tube. For example, damage to the middle ear, which carries the task of transmitting sound energy effectively, or the Eustachian tube, which sustains equal air pressure between the midear cavity and the external canal, may result in mechanical deficiency of sound transmission. In pure conductive hearing loss, no damage exists in the inner ear or the neural system. Conductive hearing losses are generally treatable.

Sensorineural deafness, which is a far more accurate term than the ambiguous terms 'nerve deafness' and perceptive deafness, describes the effect damage that lies medial to the stapedial footplate---in the inner ear, the auditory nerve, or both. In the majority of cases sensorineural deafness is not curable. The term 'sensory hearing loss' is applied when the damage is localized in the inner ear. Applicable synonyms are 'cochlear' or'inner-ear' hearing loss. "Neural' hearing loss is the proper terminology to describe the result of damage in the auditory nerve proper, anywhere between its fibers at the base of the hair cells and the auditory nuclei. This category also encompasses the bipolar ganglion of the eighth cranial nerve.

Mixed hearing loss results from conductive hearing loss accompanied by a sensory or a neural (or a sensorineural) in the same ear. Otologic surgery may help in cases of mixed hearing loss in which the loss is primarily conductive accompanied by some sensorineural damage to a lesser degree.

Functional hearing loss, which occurs far less frequently than the hear loss type described above and presents a greater diagnostic challenge in clinics, constitutes the condition in which the patient does not seem to hear or to respond, yet the handicap cannot be attributable to any organic pathology in the peripheral or the central auditory pathways. The basis for this type of hearing difficulty may be caused by entirely emotional or psychological etiology. Psychiatric or psychological therapy may be called for, rather than otological treatment.

Central hearing loss, or *central dysacusis,* remains mystifying to otologists, although information about this type of hearing defect is accumulating.

Patients suffering this type of condition cannot interpret or understand what is being said, and the cause of the difficulty does not lie in the peripheral mechanism but somewhere in the central nervous system. In central hearing loss, the problem is not a lowered pure-tone threshold but the patient's ability to interpret what he or she hears. It is obviously a more complex task to interpret speech than to respond to pure-tone signals; consequently, the tests necessary to diagnose central hearing impairment must stress measuring the patient's ability to process complex auditory information. It requires an extremely skilled, intuitive otologist to make an accurate diagnosis.

Characteristics of Hearing

If the sound is audible, the amplitude of the sound is said to be above *threshold;* and if the sound is inaudible, the amplitude is considered to be *below threshold.* The amplitude of the sound at the transition point between audibility and inaudibility is defined as the *threshold of hearing.* Once sound amplitude exceeds threshold, the sound is processed and perceived as having certain qualities including loudness, pitch, and a variety of other perceptive traits such as information. The study of auditory perception in relation to the physical characteristics of sound defines the field of *psychoacoustics.*

Sensitivity. The ear is not equally sensitive to all frequencies. The absolute sensitivity of the ear, defined by its threshold, depends on a variety of factors, the most important of which is the sound pressure level and the frequency of the sound. The resonance of the ear canal, the level effect of the ossicles, and the difference between the surface area of the eardrum and of the stapes footplate all affect the intensity of the sound that actually penetrates the cochlea.

An *audiometer* which generates signals of varying frequency and intensity is used to measure an individual's hearing sensitivity. The signals produced by the audiometer can be directed either to earphones or to a loudspeaker in an anechoic chamber. As it is far more difficult to ascertain the intensity of the sound at the level of the cochlea, and such a determination would not accurately represent how an individual hears under normal circumstances. we generally specify hearing sensitivity in terms of thresholds for sounds of various frequencies of which soundpressure levels were determined in a sound field without the listener present. Figure 10-5 maps the hearing sensitivity of the normal young human ear over a range of frequencies. The solid curve, referred to as the *minimum audible field,* or *MAF* describes the minimum intensities that can be detected when the listener is situated before a loudspeaker at a prescribed distance. Both of the listener's ears are stimulated simultaneously by the sound source, i.e. the loudspeaker.

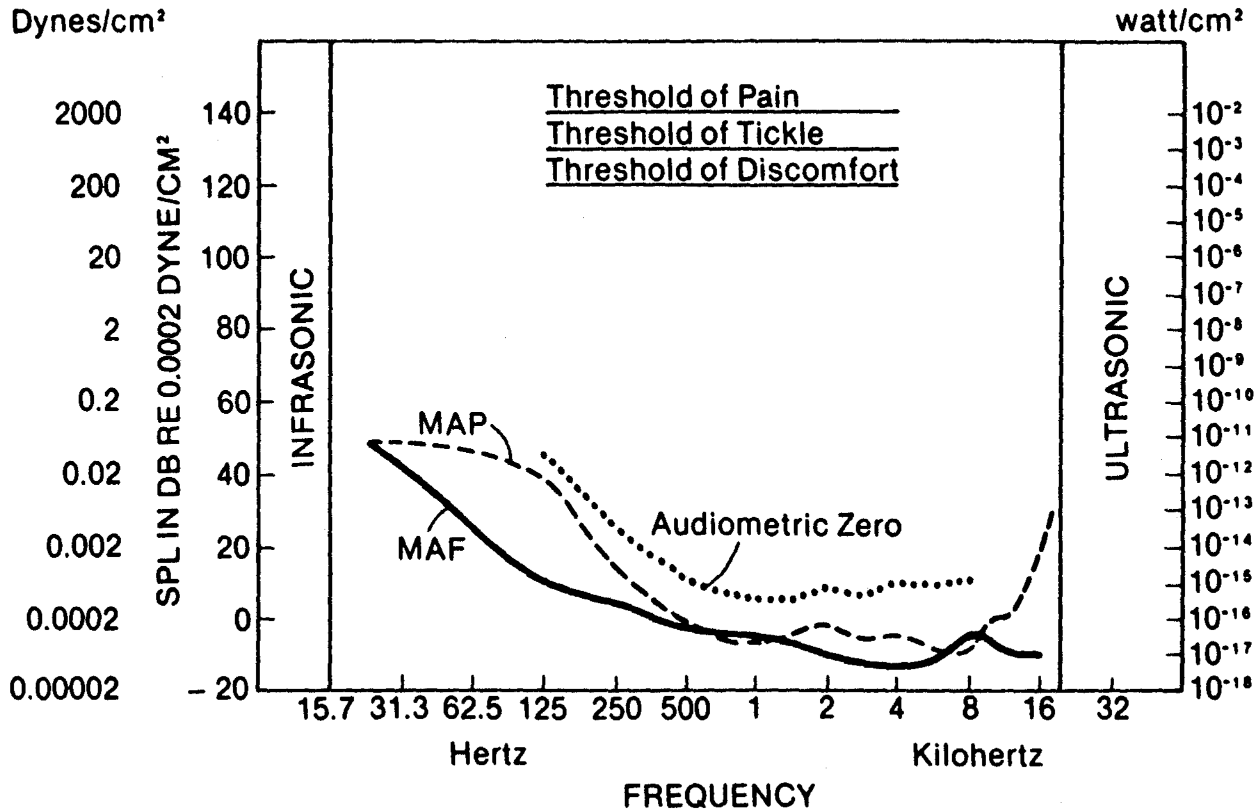
However, most clinical work in audiology entails measurements in reference to a single ear rather than to both ears. This is usually performed by directing the test signals to the appropriate earphone of a headset rather than to a loudspeaker. The use of a headset as opposed to exposure to a loudspeaker considerably modifies the listening situation. For example, the resonant frequency of the ear canal is shifted because both ends of the canal are sealed in contrast to the situation when the canal is open to the sound field. Moreover, the placement of the earphones may give rise to unwanted physiological noise that can interfere with the detection of low-frequency sounds. Also, the method of calibrating sound from a loudspeaker differs from that for calibrating sound from an earphone. Because of these differentiating and other factors, the measurement of thresholds through the use of headphones is called minimum *audible pressure,* or MAP. The MAP measurements are contrasted with MAF measurements in Figure 10-5. The threshold curve for the MAP condition appears to be several dB higher (i.e. showing lower sensitivity) than the MAF curve, a situation referred to as the 'missing 6 dB". This can be attributed to the fact that using both ears in a sound field enhances sensitivity, in contrast to listening with only one ear under an earphone; and other factors occur such as the diffraction of sound around the head in a sound field, the different resonances of the external ear canal, etc.

The two curves of Figure 10-5 represent thresholds that are two standard deviations below the mean, i.e. the curves represent the thresholds of approximately 2.5% of young adults (16-25 years of age) determined by examination to be otologically normal. These curves are based on data given in two separate studies performed four years apart conducted by the National Physical Laboratory in Great Britain. Tables 10-1 and 10-2 list the mean and standard deviations reported in these two studies and the data points (two standard deviations below the means) on which the MAP and MAF curves are based.

The intensities defining an audiometric zero at each of the standard frequencies on a pure-tone audiometer are represented by the dotted curve of Figure 10-5. These intensity values were established by international agreement among scientists as being representative of the average minimum audible soundpressure levels for young adult ears and have been incorporated in the standards for audiometric calibration throughout most of the world. This standard. as the result of being adopted in 1969 by the American National Standards Institute, is referred to as ANSI-1969 standard and it is listed with the MAP values in Table 10-1.

Loudness. While the sensation of loudness correlates to the amplitude of the sound above thresholds, loudness does not become manifest to the human ear in equal measures as the amplitude increased over different frequencies above the threshold. Individual judgment constitutes the deciding factor in ascertaining the degree of loudness. This had led to the development of equal loudness contours which are curves connecting SPL points of equal loudness for a number of frequencies, as judged by tested listeners. These curves, also called *phon*[2] *curves*, are constructed by asking subjects to judge when tones of various frequencies are considered equal in loudness to a 1-kHz of a given SPL. The official definition (ANSI, 1973) of the phon specifies binaural (two-ear) listening to the stimuli

[2]A phon is a unit of loudness that, at the reference frequency of 1 kHz, is equated to the decibel scale.



10-5

Figure [redacted] The area of human audibility. The two lower curves represent the lowest (best) thresholds of hearing of young adults. The solid curve is the minimum audible field (MAF) and the dashed curve is minimum audible pressure (MAP). (See text for an explanation of these terms). The dotted curve is the current standard for audiometric zero (From ANSI. 1969. *American National Standard Specification for Audiometers.* (ANSI S3.6-1969, revision 1989). American National Standards Institute. New York: Acoustical Society of America.) The upper three curves represent averages for sensations of discomfort, tickle, and pain. The ordinates define intensity in terms of pressure in dynes/cm², sound-pressure level in dB, and power flow in watt/cm².

in a sound field. Equal-loudness contour curves are given in steps of 10 phons in Figure *10-6,* with the dashed MAF curve from Table 10-2 included in the plot as a threshold reference.

As an example of how humans perceive sound, consider a 30 Hz tone at *95* dB SPL. It would be judged by a typical listener as being as equally loud as a 1000 Hz tone at 70 dB SPL or a *5000-Hz* tone at *65* dB SPL As sound is steadily increased in intensity above the threshold, it will eventually cause the listener to experience physiological discomfort. A further increase in the intensity produces a tickling sensation in the ear; and an additional increase in intensity causes the listener to experience pain. These three levels constitute, respectively the *thresholds of discomfort, tickle and pain,* which are represented by the upper three lines of Figure 10-5. While these threshold values of 120, 130 and 140 dB SPL represent statistical averages for young adult ears, different individuals have different tolerance thresholds; but these values do not differ markedly from the statistical averages. Thus, in Figure 10-5, the region between the discomfort and the audiometric zero constitute the usable dynamic range of hearing for humans.

Pitch. The sensation of pitch is obviously related to the frequency of the tone. The actual pitch of a sound is affected by other factors, including the sound pressure level and the presence of component frequencies. Pitch perception is a complex process, one that is not yet fully understood. Pitch elicited by some sounds may evoke the same aural response whether or not the fundamental frequency is present. The average adult male voice carries a fundamental between

10-1

TABLE ~~9-5~~ The means and standard deviations and the data points (two standard deviations below the means) on which the MAP curves of Figure ~~9-32~~ 10-5 were based.

FREQUENCY IN Hz & kHz	MEAN SPL IN dB	σ IN dB	SPL AT 2 σ BELOW MEAN	ANSI-1969
80 Hz	61.0	8.0	45.0	
125	45.5	6.8	31.9	45.5
250	28.0	7.3	13.4	24.5
500	12.5	6.5	−0.5	11.0
1 kHz	5.5	5.7	−5.9	6.5
1.5	8.5	6.1	−3.7	6.5
2	10.5	6.1	−1.7	8.5
3	7.0	5.9	−4.8	7.5
4	9.5	6.9	−4.3	9.0
6	10.5	9.1	−7.7	8.0
8	9.0	8.7	−8.4	9.5
10	17.0	9.0	−1.0	
12	20.5	9.6	1.3	
15	39.0	10.7	17.6	
18	74.0	21.9*	30.2	

*Calculated from reported standard error of the mean.

Data on mean sound-pressure levels and standard deviations (σ) for 80 through 15 kHz taken from R. S. Dadson and J. H. King, "A Determination of the Normal Threshold of Hearing and Its Relation to the Standardization of Audiometers," *Journal of Laryngology and Otology* 66 (1952): 366–78, as reproduced in *Forty Germinal Papers in Human Hearing*, ed. J. Donald Harris (Groton, CT: The Journal of Auditory Research, 1969), pp. 48–58. The data for 18 kHz were taken from J. Donald Harris and C. K. Myers, "Tentative Audiometric Hearing Threshold Level Standards from 8 through 18 Kilohertz," *Journal of the Acoustical Society of America* 49 (February 1971): 600–601.

10-2

TABLE ~~9-5~~ The means and standard deviations and the data points (two standard deviations below the means) on which the MAF curves of Figure ~~9-32~~ 10-5 were based.

FREQUENCY IN Hz & kHz	MEAN SPL IN dB	σ IN dB	SPL at 2 σ BELOW MEAN
25 Hz	63.5	8.0	47.5
50	43.0	6.5	30.0
100	25.0	5.0	15.0
200	15.0	4.5	6.0
500	5.5	4.5	−3.5
1 kHz	4.5	4.5	−4.5
2	0.5	5.0	−9.5
3	−1.5	6.0	−13.5
4	−5.0	8.0	−21.0
6	4.5	8.5	−12.5
8	13.5	8.5	−3.5
10	16.5	11.5	−6.5
12	13.0	11.5	−10.0
15	24.5	17.0	−9.5

Data on mean sound-pressure levels and standard deviations (σ) taken from D. W. Robinson and R. S. Dadson, "A Re-determination of the Equal-Loudness Relations for Pure Tones," *British Journal of Applied Physics* 7 (1956): 166–81, as reproduced in *Forty Germinal Papers in Human Hearing*, ed. J. Donald Harris (Groton, CT: The Journal of Auditory Research, 1969), pp. 185–200.

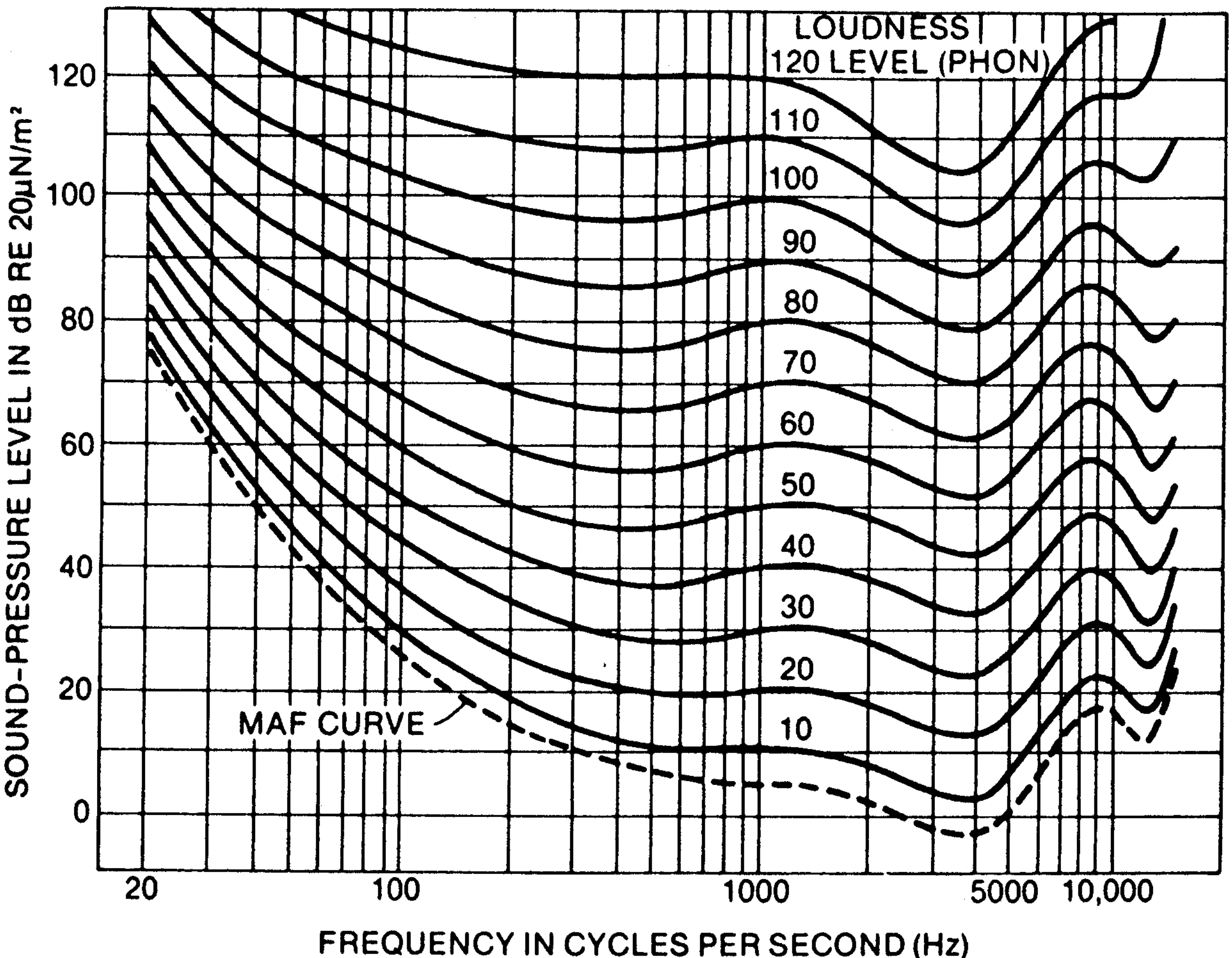


Figure 10-6 Equal-loudness contours. (From A. P. G. Peterson and E. E. Gross, *Handbook of Noise Measurement*, 1980, General Radio. publisher.)

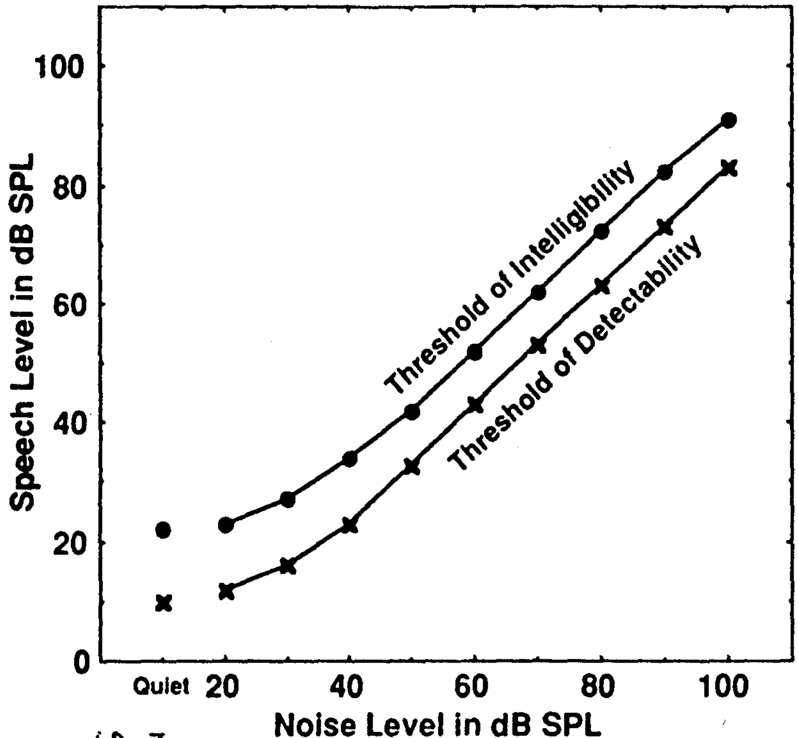


Figure 10-7 The effect of white noise on the thresholds of detectability and intelligibility of running speech. (Adapted from Hawkins, J. E., Jr., and Stevens, S. S. 1950. "The masking of pure tones and of speech by white noise." *Journal of the Acoustical Society of America,* 22, 12.)

120 and 150 Hz and that of atypical adult female lies between 210 and 240 Hz. Yet we generally find it easy to distinguish between male and female voices even though the telephones does not transmit frequencies much lower than 300 Hz. Somehow we are able to atone aurally for the fundamental frequency missing from the signal passing through the telephone receiver.

The spectrum of a sound generates a psychological sensation of quality. This permits us to distinguish the difference, say between a trumpet and an English horn playing the same note because of the differences in their respective sound spectra that are in turn are functions of the complexity of their vibrations and the resonance modes inherent in their respective constructions. We are also able to discern different speech sounds because of the differences in the sound spectra. Even over the telephone, individual voices are recognizable because of the differences in their sound spectra.

Masking. Masking is said to have occurred when the audibility of a sound is interfered with by the presence of noise or other background sound. The "cocktail party" effect, which makes it difficult to carry on a private conversation against a backdrop of other people's chatter, is a prime example of masking. Speech becomes unintelligible by the presence of excessive background noise. Although masking is almost always undesirable, broadband noise may be introduced into an office environment to make conversation unintelligible to an adjacent office. Since most of the intelligence in speech is generally contained in the frequency range between 200 Hz and 6 kHz, noise in that frequency range is most objectionable in terms of speech masking. But excessively loud noise in any frequency band can adversely affect speech intelligibility by causing an overload of the auditory system that it cannot effectively discriminate speech from the prevailing total signal. Consonants essential to conveying verbal information tend to be pronounced softly, so they become readily indiscernible in the presence of noise.

A person speaking normally produced an unweighted sound level of 55 to 70 dB at 1 m. It is more taxing for that person to speak more loudly for a sustained time. A typical maximum voice effort, in the form of a shout, produces about 90 dB at 1 m. Speech intelligibility generally improves when the speaker and the listener are near one another and if the speaker increases the signal-tonoise ratio by talking louder. Maximum intelligibility usually can be obtained if the unweighted level of the speech is between 50 and 75 dB at 1 m from the speaker. Speaking more loudly does not always guarantee greater intelligibility, even though the signal-to-noise ratio (defined as the intensity of the signal divided by the intensity of the noise) may be increased, since the formation of speech sound above 75 dB may degrade sufficiently that there is little or no improvement in intelligibility. If a listener is familiar with the words and the dialect used, intelligibility will be greater. It is for this reason that critical communications, particularly those of air controllers, are based on a limited vocabulary. In ordinary face-to-face conversation, the listener has the additional luxury of making out the context of the words by observing the speaker's facial expressions and gestures.

Prediction of Speech Intelligibilitv---The Articulation Index

In order to assess the effect of noise on speech communication, it is necessary to conduct speech-intelligibility tests with actual speakers and listeners in the presence of interfering noise. The test materials may be sentences, digits, disyllabic words, monosyllabic words, or nonsense syllables. The listeners are scored according to the percentage of the speech materials heard correctly. The background interfering noise is generally recorded and played back in the testing laboratory. From such experiments came the realization that speech intelligibility is a function of the intensity and the frequency characteristics of the interfering noise. Regarding the S/N ratio, Licklider and Miller stated that the S/B should exceed 6 dB for satisfactory communication, although the presence of speech may be detected for S/N as low as -18 dB. If the intensity of the signal (speech) exceeds the noise, the sign of the S/N value is plus; conversely, a negative value of S/N indicates that the noise is more intense than the signal. Figure 10-7 maps the effect of white noise on the thresholds of detection and intelligibility of running speech. According to this figure, which was developed by Hawkins and Stevens based on extensive tests making use of running speech and white noise, the threshold of intelligibility occurs when the level of the speech exceeds the noise level by about 6 dB (S/N of 6 dB). As the sound pressure level of the noise is increased above this value, the threshold of intelligibility is proportionally increased so that the S/N value of -6 dB remains fairly constant over a wide range of intensities. For other kinds of speech materials and different masking noises, the relationship between the threshold of intelligibility and the level of the interfering noise may not necessarily remain the same. At a S/N of -18 dB running speech can be detected but not understandable.

Other but simpler methods have been developed for measuring the effect of interfering noise on the intelligibility of speech. A principal method of predicting speech intelligibility is the articulation *index*, or AI, which is a value that ranges from 0.0 to 1.0 and represents the proportion of the speech spectrum that occurs above the noise. French and Steinberg of the Bell Laboratories developed the concept of articulation index on the basis of the assumption that most of the intelligence in speech is contained in the frequency bands between 200 and 6100 Hz. The articulation index can be calculated from the levels of the masking signal and the speech level in the frequency bands. The contribution of each frequency band to speech intelligibility is defined as 12 dB plus the sound level of the speech less the masking level. Each frequency band's contribution is limited to the range between 0 and 30 dB. The sound level of the speech signal is based on a long term energy average in each frequency band; and each frequency band contribution is multiplied by a weighting factor. The sum of the weighted contributions divided by 10,000 yields the articulation index.

Table 10-3, based on the division of the speech spectrum into one-third octave, provides the data necessary to calculate articulation index. The first column lists the center frequency of the one-third octave, the second column gives the typical male voice long-term average speech level plus 12 dB at 1-m distance. The weighting factor for each one-third octave band is listed in the third column.

Center Frequency (Hz)	Speech Level (+12 dB)	Weighting Factor
200	67	4
250	68	10
315	69	10
400	70	14
500	68	14
630	66	20
800	65	20
1000	64	24
1250	62	30
1600	60	37
2000	59	37
2500	57	34
3150	55	34
4000	53	24
5000	51	20

Table 10-3 Weighting factors for one-third octave band-based calculation of articulation indexes. Speech levels given in the second column are those for a typical male voice at 1 m.

If the masking noise has been measured only in full octave bands, Table 10-4 may be used instead of Table 10-3.

Center Frequency (Hz)	Speech Level (+12 dB)	Weighting Factor
250	72	18
500	73	50
1000	78	75
2000	63	107
4000	58	83

Table 10-4. Weighting factors for one octave band-based calculation of articulation indexes. This is for the typical male voice level at 1 m.

Example Problem, Calculation of AI:

Let us compute the articulation index of a male voice speaking at a normal level 1 m from the listener in the presence of pink noise that contributes 47 dB in each octave band. Table 10-3 is used for this calculation. The 47 dB octave-band noise level is subtracted from the values in the second column; and the difference (up to a maximum of 30 dB) is multiplied by the weighting factors of the third column.. The resulting weighted contributions are added and divided by 10000, yielding the articulation index. Table 10-5 below gives details of the calculations, with the values given in the fifth column produced by subtracting 47 dB from the speech level of the second column and each of the values in the fifth column is multiplied by the weighting factor of the third column to yield the figures listed in the sixth column.

Center Frequency (Hz)	Speech Level (SL) (+12 dB)	Weighting Factor (WF)	Noise (N)	SL-N= DIFF	WF x DIFF
250	72	18	47	25	450
500	73	50	47	26	1300
1000	78	75	47	303	2250
2000	63	107	47	16	1712
4000	58	83	47	11	913

Articulation index = (450+1300+2250+1712+913)/10000 = 0.6625

Table 10-5. Calculations for the articulation index in the sample problem.

The articulation index is 0.6625, Or 66.25%.

Speech Interference Level (SIL)

Measurements to obtain data for articulation indexes require special laboratory equipment for determination of S/N in a number of frequency bands. A simpler procedure of estimating the effect of noise on verbal communication makes use of octave-band levels as measured in a typical noise survey. The parameter that is called the speech *interference level,* abbreviated SIL, can be obtained by

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[3]The value of (Speech Level + 12 dB - Noise Level) must fall between 0 and 30.

computing the arithmetic average of octave band levels in the three octave bands of 600-1200, 1200-2400, and 2400-4800 Hz. However, current practice uses the arithmetic level in the 'preferred' octave bands with center frequencies at 500, 1000 and 2000 Hz. Speech interference level defined thusly is referred to as PSIL. The speech interference level PSIL = 68 dB has been identified as the level at which reliable speech communication is barely possible in a normal male voice at a distance of 0.3 m (or 1 ft) outdoors. If a male speaker talks in a raised voice, a very loud voice, or shouts, the speech interference levels have been identified respectively as PSII, = 74, 80, and 86 dB A female speaker, on the average, has PSIL levels 5 dB less than the corresponding values for a male. Table 10-6 lists the PSIL, (in dB) at which effective speech communication is barely possible.

PSIL (dB)

VOICE EFFORT

Distance	Normal		Raised		Very Loud		Shouting	
(m)	M	F	M	F	M	F	M	F
0.3	68	63	74	69	80	75	86	81
1	58	53	64	59	70	65	76	71
2	52	47	58	53	64	59	70	65
3	48	43	54	49	60	55	66	61
4	46	41	52	47	58	53	64	59

Table 10-6. PSIL (in dB) at which effective speech communication is barely possible.

The table is based on minimally reliable communication, in which about 60% of the communication of uttered numbers and words out of context can be discerned. In order to roughly approximate PSIL in terms of dBA, 7 dB can be added to the values of PSIL.

Example Problem, SIL:

Background noise levels for an industrial plant were measured to be 62, 65, and 74 dB, respectively in the 500-, 1000-, and 2000-Hz center frequency bands. What are the implications for speech interference at a distance between a speaker and a listener standing 1 m apart?

To solve this problem, the arithmetic average of the noise level in three bands are first determined. This will be (62 + 65 + 74)/3 = 67 dB From table 106 we establish that reliable speech is barely possible for a male speaking in a raised voice or a female speaking in a very loud voice.

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Designing Circuits for Hearing Instruments

Ulrich Kleine, Manfred Mauthe

A second technological revolution is changing hearing instrument technology. The recent development of VLSI processes with smaller feature sizes and lower supply voltages has allowed researchers to implement increasingly sophisticated signal processing algorithms that allow users to more clearly hear speech while attenuating background noise. In addition, advances in programmability have opened the door to self-adjusting hearing instruments.

The invention of transistors 30 years ago facilitated the first linear BTE (behind-the-ear) and ITE (in-the-ear) hearing instruments. These devices made it possible for the first time to successfully correct hearing loss caused by conductive deafness or presbycusis – conditions in which the threshold of pain is raised by as much as that of audibility.

But in recent times, modern submicrometer processes with finer structures and lower supply voltages have enabled the integration/implementation of more and more complicated signal processing algorithms. Furthermore, thanks to increasing programmability, it is now possible to fit hearing instruments to the wearer's residual requirements. These developments have, in turn, posed new challenges in terms of diagnostic procedures, algorithm development, and hearing instrument hardware.

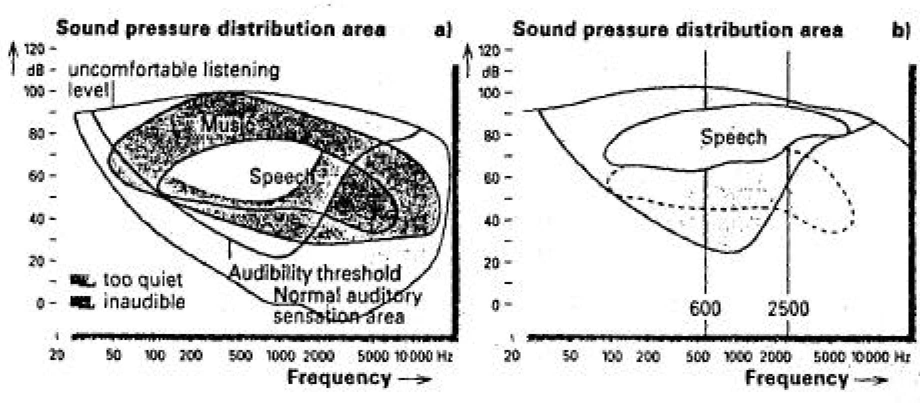
An Answer to the Loss of Speech Audibility

Approximately 24% of individuals between 15 and 75 years of age are hearing impaired. This makes impaired hearing, which is often caused by infectious diseases or noise, as common as poor vision. Often, the primary problem is a diminished ability to understand speech. Hearing instruments can help by adapting the dynamic range of speech to the dynamic range of the patient while suppressing background noise. The annoying whistling that often occurs with hearing instruments is caused by acoustic feedback when the gain is set too high. One way of preventing this is to use an adaptive filter.

Since there are many different kinds of hearing impairment, all of which cannot be dealt with in this paper, we have restricted ourselves here to one commonly occurring form of hearing loss. Fig. 1 shows the threshold of audibility in a sound pressure versus frequency diagram for a person with normal hearing, the so-called "auditory sensation area" (blue shaded). A typical hearing impairment caused by labyrinthine deafness is also shown in the figure. The threshold of audibility is raised but the uncomfortable listening level is scarcely changed. The figure indicates which sounds are inaudible. The individual, frequency-dependent threshold of audibility is the sound pressure level at which a sound can be perceived. The upper limit of the dynamic range is defined by the pain threshold. In hearing instrument diagnostics, the slightly lower, uncomfortable listening threshold is used for the same purpose. It is also typical that the threshold varies considerably with frequency.

Ulrich Kleine

Since 1985 Dr. Kleine (40) has been with Siemens' Corporate Research and Development Department, working in the field of analog and digital signal processing. Trained as an electronics engineer, Dr. Kleine wrote his dissertation on the subject of "Integrated Circuit Compensator Filters in CMOS Technology".



In order to compensate for such hearing impairments in an acceptable way, multi-channel, non-linear signal processing is required. Figure 1 also shows how the auditory sensation area of speech is adapted dynamically to limited hearing ability with three independently working compression amplifiers in the three frequency channels. In this way, a high level of speech comprehension is obtained by maximum utilization of the residual hearing ability.

In addition to correcting amplitude response, the frequency response and temporal behavior of speech also play an important role in speech comprehension. In order to correct or at least alleviate hearing problems in these areas, a programmable, three-channel hearing instrument circuit using CMOS low-voltage technology has been developed in cooperation with Siemens Audiological Engineering GmbH (S.A.T.), at the company's Corporate R&D Department.

Low Voltage Technology for Hearing Instruments

In addition to achieving a satisfactory processing rate, the greatest challenge for circuit design techniques is presented by the low power dissipation required by hearing instruments. For both analog and digital CMOS circuits, power dissipation is approximately proportional to the operating frequency, the square of the supply voltage, and the load capacitances. In analog circuits, the operational speed is determined by the system specifications; in analog sampled-data circuits it is determined by the system clock. Noise is directly proportional to absolute temperature and inversely proportional to the loading capacitances. Thus the capacitances are defined by the noise specifications. For this reason, reduced power dissipation can only be obtained with reduced supply voltage. In addition, small volume is a secondary prerequisite for hearing instruments, which results in the battery being limited to a nominal 1.2 or 1.6 V cell.

Manfred Mauthe

With Siemens since 1974, Mr. Mauthe (45) worked in the area of charge coupled devices for image sensor and filter applications before turning his attention to the development of low power CMOS circuits for programmable hearing instruments. Since 1992, he has been active in the Microelectronics Applications Center of the company's Corporate Research and Development Department.

The extensive signal processing requirements of hearing instruments, e.g. programmable filtering and volume compression as well as adjustable gain control, can be implemented using so-called switch-capacitor (SC) circuits. In these circuits, charge packets serve as signal variables and the time constants are set via capacitance ratios and clock frequencies. Here, a CMOS process is preferred as compared with a pure bipolar one, since with MOS transistors switches are simple to implement, charges can be read non-destructively, and the entire supply voltage range can be retained as the output swing due to the complementary transistors. In addition, digital memories and registers can be manufactured on smaller areas and with lower power requirements using MOS transistors rather than bipolar ones. The starting point for this development was a 3 µm CMOS process with analog capability for supply voltages of 5 V and threshold voltages between 0.7 and 0.9 V. In order to be able to operate part of the circuitry at a cell voltage of only 1.2 V, the threshold voltages had to be reduced to 0.4–0.6 V. In this way, CMOS amplifiers with a 2 V supply voltage and an output swing of 0.5 V became achievable. A further reduction of the threshold current was not possible due to substantially increased reverse currents in the transistors. Recently, suggestions have been made for achieving a further attenuation in the threshold voltage using substrate bias voltages or split-gate technology. Under this scenario the active transistors have a lower threshold voltage while the switching transistors have a higher one.

In order to achieve a voltage of more than 2 V with just one cell, a doubling of voltage is required. Usually, voltage multipliers are based on diodes. However, since the diode reverse voltage is approximately half that of the supply voltage, a cell voltage of 1.2 V only achieves a small increase in voltage. With switched transistors this effect can be avoided, since their saturation voltages are less than diode threshold voltages. By separately

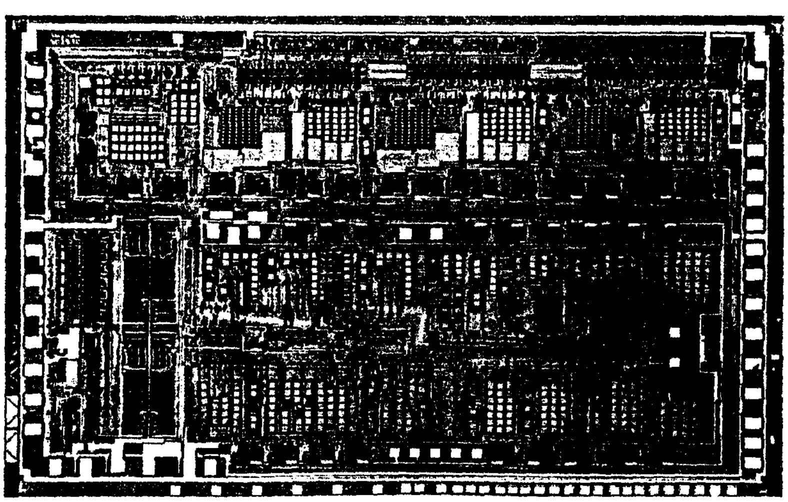
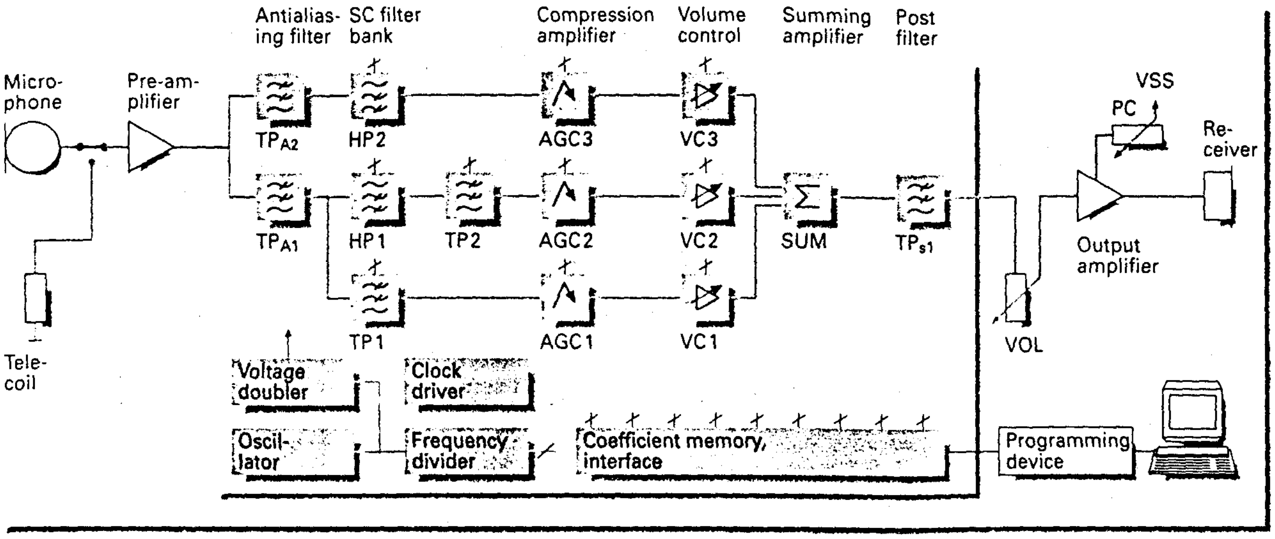


Figure 2
A programmable three-channel hearing instrument circuit with over 3000 transistors on a chip area of 33 mm²



controlling the CMOS inverter transistors, a cross current can be avoided during switchover, thus avoiding unnecessary energy consumption.

Just as is the case with analog circuits, the main method for reducing power dissipation in digital circuits is by cutting back the supply voltage. For standard CMOS gates, the noise ratio is so great that in standard processes, supply voltages of 1 V can still be used. The advantage of digital as opposed to analog circuits is that data can be stored for as long as required, without errors. With a view to the adaptive hearing instruments of the future, this is of great importance.

On the other hand, analog circuits exhibit less power dissipation with moderate signal-noise specifications and are characterized by simpler signal processing.

Toward a Programmable Circuit for Analog Hearing Instruments

In order to meet auditory requirements, a three-channel circuit with a programmable compression amplifier in each channel was developed (Fig. 2). The circuit is implemented 3 µm CMOS low-voltage technology with a high-ohm p-well and two polysilicon layers, and has a chip area of 33 mm^2, including bump pads. In addition to numerous passive elements, it contains more than 3000 transistors. Its power requirement is 1.3 mW for a voltage of 1.3 V. The circuit is part of a supply modular hearing instrument concept. It is switched between the pre-amplifier and output amplifier of a conventional hearing instrument (Fig. 3).

The amplified input signal is at first band-limited to the Nyquist range with a pre-filter, since subsequent filtering is carried out in discrete time. Noise minimization is achieved by using two filters with corner frequencies of 5 and 15 kHz. Finally, the output signals from the pre-filter are again amplified since the succeeding filter bank permits a greater degree of control. The split input signal is divided into low-pass, band-pass and high-pass channels by means of an SC filter bank.

Due to their limited sensitivity to deviations in component values, SC filters that imitate two-way resistive closed LC reference filters were selected for the channel filter. These filters also have little tolerance sensitivity with regard to variations of their amplifiers in the open-loop

Figure 3
Block diagram of the TRITON 3000 three-channel hearing instrument

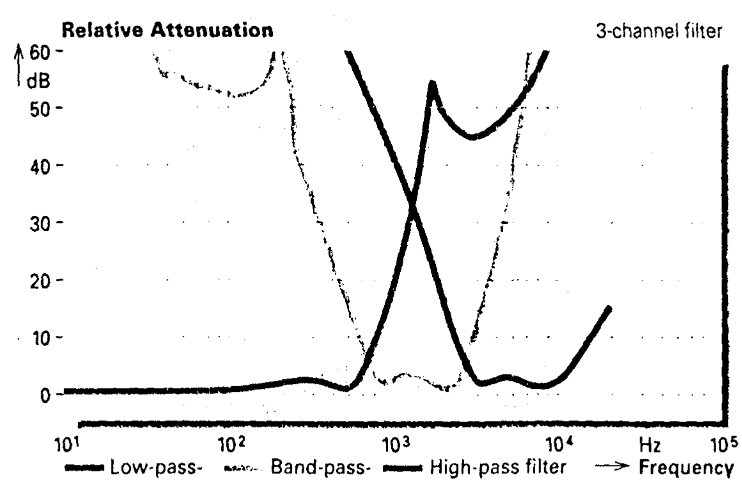
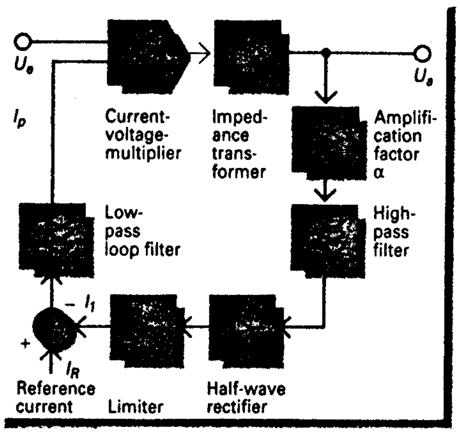


Figure 4
Frequency responses of the filters in the programmable three-channel hearing instrument circuit



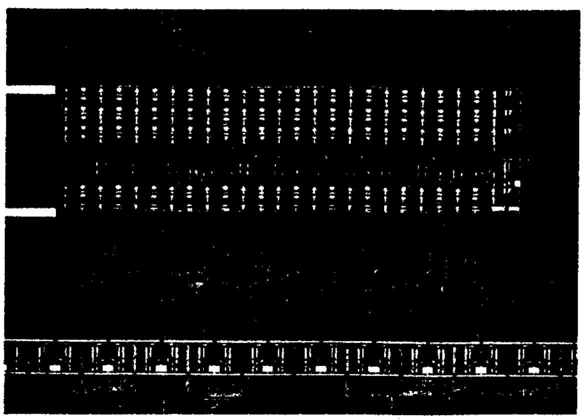


Figure 5
Block diagram of the compression amplifier in the programmable three-channel hearing instrument circuit

Figure 6
Test circuit used to investigate the power dissipation, delay time and noise immunity of a potential digital hearing instrument

gain. This is especially significant for hearing instruments, since the operating voltage with a weakening battery can change drastically. Single-stage transconductance amplifiers with cascade outputs serve as amplifiers.

The SC low-pass is a stray-insensitive, elliptic leapfrog of the 3rd order. By sampling during both clock phases, the Nyquist range is doubled and an additional pole position created at the Nyquist frequency. Although the low-pass filters were designed with a hyperbolic sine transformation, their transfer function resembles the frequency characteristic of a filter designed with a bilinear transformation, due to this modification. Pure SC high-pass filters with steep slopes tend to be unstable. Since they usually have internal nodes that do not have a direct voltage path to ground, charge accumulations can be caused by leakage current charges and reverse currents; this leads to the breakdown of the circuits. For this reason, a single-side band-pass was selected instead of a high-pass and was implemented with a bilinear leapfrog structure with coupled SC biquads. The use of band-passes instead of high-passes supports the post filter, which is designed to dampen frequencies above 10 kHz. In the TP1 and HP1 filters the capacitances can be switched over in two stages; and in the TP2 and HP2 filters in four stages. Figure 4 shows the frequency responses of the three channel filters.

In each channel, the speech components can be compressed with a compression amplifier. The amplitude-dependent gain of the compression amplifiers is equal to 1 up to an adjustable threshold. Above this threshold it decreases. In contrast to conventional AGC (automatic gain control) circuits, the increase remains finite so that the user can also differentiate between levels of volume. The threshold that arises during compression can be set in six steps of 6 dB each. Apart from the filter bank, these are the most important and most elaborately designed circuit blocks (Fig. 5). The feedback circuit consists of a current-voltage-multiplier, an SC isolating amplifier for impedance matching, a programmable SC amplifier for setting various compression characteristic curves, a loop high-pass, a half-wave rectifier, and a current differential circuit for producing control current for the multiplier. For multiplication, a resistively loaded transconductance amplifier is used whose input differential stage is operated in weak inversion and whose slope is therefore linearly dependent on the control current.

Finally, each of the signals passes through an offset compensated, 5 bit binary loaded, programmable SC amplifier in which the amplification factor can be set in 2 dB steps. In a summing amplifier, the three speech components are reunited in proper phase. Finally, a post filter deletes the high frequency noise just ahead of the output stage.

Coming Up: Digital Signal Processing

In order to attain more flexibility in correcting individual hearing impairments, we are investigating how far existing CMOS-based digital signal processing can go in improving the quality of hearing instruments. In principle, digital signal processing has the advantage of simple data storage and the potential of offering a simple and efficient realization of signal-dependent branches within the signal flow. Adaptive algorithms, in particular, can only be calculated digitally.

Using a test circuit (Figure 6), the power dissipation, delay time, and noise immunity of digital circuits with supply voltages of 1 to 1.5 V were investigated. The test circuit comprised four registers, a multiplexer and a carry-select- adder-subtracter. This adder was characterized by the logarithmic dependence of the delay time on word size, similar to carry-look-ahead-adders. However, the carry-select-adder-subtracters have a regular layout. Simple carry-ripple-adders, which facilitate an even more compact layout, are linearly dependent on word size. The circuit was implemented in 0.6 µm CMOS technology and the 24-bit wide data path was set up using the bit-slice technique. The total circuit contained 7,924 transistors on an active surface of only 0.46 mm^2. The circuit was fully functional for supply voltages down to 0.9 V. The power dissipation measured was 626 µW for a supply voltage of 1.5 V and a clock frequency of 10 MHz; it was 121 µW for a supply voltage of 1 V and a clock frequency of 5 MHz.

The results obtained from the test circuit show that digital hearing aids with power dissipation of less than 2 mW can be implemented with current technology. In addition to the eight different setting variables of the three-channel hearing instrument circuit, further parameters can also be varied. Thus we are clearly one step closer to attaining the goal of an individually programmable hearing instrument that optimally adjusts itself to suit various listening environments. •

Prepared by Ron Adrezin Adjunct Assistant Professor of Mechanical Engineering, The Cooper Union Gateway Project August 26, 1994 Project: Wheelchair Seating and Mobility

DESCRIPTIONS OF ITEMS APPEARING IN CHART 1

The following is the rationale for including the below items as modules in an interdisciplinary rehabilitation engineering course. The focus of these modules is the development of knowledge for seating and mobility for wheelchair users. It is desired to safely maximize the user's function, comfort and independence. These modules will be integrated with those prepared by Dan Bogen (U. of Penn.) for wheelchair design.

Ph ysiology - the physiology and anatomy of the human body Rationale: Inclusion of this module on the function of the human body will form a basis for the understanding of physical impairment and pathology. This module can be supplemented with an anatomy CD-ROM.

Pathology - the nature of the disease Rationale: Wheelchair users are often described by their pathology, for example, spinal cord injury at C4, Multiple Sclerosis (MS), Muscular Dystrophy (MD), Cerebral Palsy (CP), Amyotrophic Lateral Sclerosis (ALS, Lou Gehrig's Diesease), etc. Is the condition acute or chronic? Is the disease progressive? This information is critical before initiating a design. Would you purchase a hand control for an individual who will lose use of their upper extremity in 3 months? Can this device be activated with the remaining function? The reality of funding does not allow for the frequent acquisition of equipment and services. Funds are generally not available and there may be a 1 year or more wait for the delivery of necessary equipment and services.

Impairment - the effects of the pathology on the human body Rationale: The students will learn the result of different pathologies (e.g. diseases, trauma) in terms of diminished or lost function (e.g. spasticity, muscle atrophy, paralysis, limb-loss).

Disabili - the inability to perform specific activities (e.g. walking, hearing, seeing, writing) Rationale: Rehabilitation engineers generally design to modify an individual's environment to accommodate their disability. When designing a ramp for the public, we are concerned for individuals who may not climb stairs. Whether they are confined to a wheelchair due to impairments that include paralysis or limb-loss or pathologies such as ALS or spinal cord injury does not impact our design. However, when designing for an individual, we must look at the whole person. We select technology that they will feel comfortable with and use. If the condition is progressive, the system must be flexible to allow modification as necessary with minimal expense. Examples of equipment to limit the impact of a disability include wheelchairs, walkers, hearing aids, Braille printers, automatic doors, environmental control units and television remote controls.

Handicap - the inability to perform major life roles (e.g. driving, attending school, working, cooking, toileting, parenting) Rationale: It is important for the rehabilitation engineer to distinguish between "having a handicap" and "having a disability." Two persons with identical disabilities may not necessarily have the same handicaps. An individual living in the country may be concerned with the inability to drive, the other living in the city may not. The concept of being handicapped depends on an individual's social environment. It is preferable for the person with the disability, not society, to determine whether a handicap exists.

Mechanics - the study of statics and dynamics Rationale: Modules on mechanics should be included in any interdisciplinary engineering course. In this course, it is essential for the understanding of biomechanics.

Biomechanics - forces and constraints of and applied to bones, joints and muscles Rationale: The following principles must be understood in order to design an effective seating system:

1. Multiple Loading Forces
2. Range of Motion of Joints
3. Buckling of Spine
4. Constraints
5. Pressure - Over an extended period of time, localized pressures (>2psi) can cause pressure sores. (Refer to Pressure Sore Elimination and Thermal modules.)

Thermal - the heat build-up associated with sitting for an extended period of time Rationale: High pressures and temperatures on the skin over an extended period of time result in pressure sores. The seating system must be designed to allow for adequate heat dissipation. (Refer to Pressure Sore Elimination and Pressure modules.)

Postural Control - proper positioning of an individual in a wheelchair with the aid of one or more seating components (e.g. lateral pelvic support, arm support, contoured seat and back) Rationale: It is desired to teach students to design seating and mobility systems to safely maximize the wheelchair user's function, comfort and independence. Supports may be required to position an individual with a high spinal cord injury (e.g. C4 quadriplegia) or with Cerebral Palsy. We must design to prevent, delay, or accommodate spinal and pelvic deformities. The seating and mobility system will allow, as required, the user to cook, eat, access a computer and telephone, transfer to a toilet, read and write, etc.

Stability - the stability of the user within the wheelchair, and the wheelchair on varying surfaces and inclines must be evaluated Rationale: The student must account for the friction between the individual and the wheelchair. Is there sufficient friction to holder the user in place or is a support required? This is particularly important as the wheelchair travels over rough terrain. Will the wheelchair topple as it rolls over a curb? What role does the center of gravity and wheelbase have on stability? These are examples of the application of fundamental engineering principles to rehabilitation engineering.

Pressure Sore Elimination - Pressure sores (pressure ulcers) result from the soft tissue being deprived of oxygen and nutrients. This is caused by excessive pressure, temperature and humidity. There is an inverse relationship between pressure and time in the formation of sores. A greater pressure can be tolerated for a lesser period of time. Rationale: The occurrence of pressure sores is a serious problem affecting those with limited mobility. In particular, individuals with spinal cord injuries who cannot shift their body weight and remain in one position for long periods of time are susceptible. Prevention is critical. Treatment involves relieving all pressure from the affected area for weeks. This usually results in the person remaining in bed and missing a considerable number of days of work or school. In many cases, the individual is hospitalized to prevent infection and other complications. When designing a seating system, pressure and temperature reduction is critical. This may be accomplished through passive systems such as foam or gel cushions, or with mechanical systems that shift the pressure distribution borne by the skin. (Refer to Machine Design module.)

Fluid Flotation Seats - cushions filled with air, water, viscous fluid or elastomer gel Rationale: Each medium has its own advantages and disadvantages. For example, gel provides good heat dissipation, but its pressure distribution is dependent on the seat's outer covering. Air cushions may be designed with large bladders or smaller cells to minimize any hammocking effect, but proper inflation must be maintained. These are examples of designs that would benefit from fluid mechanics and thermal dynamic analysis, instrumentation, and clinical trials.

CAD/CAM - computer-aided design/computer-aided manufacturing Rationale: CAD/CAM could enhance the design and production of custom seating. Many wheelchair users require custom seating system components and the application of CAD/CAM could both improve the results and lower the cost. (Refer to chart 2.)

Industrial Desisn - design for form and function Rationale: A device that may overcome a disability will often be left unused if it is unsightly. The stare by passerbys and the insensitivity of society is difficult for individuals with physical disabilities. Imagine strapping an ungainly device to someone and asking them to wear it in public. Any design must be unobtrusive and attractive.

Instrumentation - measurement of the pressure, temperature, humidity, etc. at the user/seat interface Rationale: Several different systems have been used to measure the user/seat interface pressure. They range from the use of multiple individual sensors to a single thin film sheet that covers the seat with a matrix of pressure sensors. This biofeedback data is then transmitted to a computer for analysis.

Machine Design - active devices for pressure sore elimination When designing a seating system, pressure and temperature reduction is critical. This may be accomplished through passive systems such as foam or gel cushions, or with active mechanical systems that reduce the pressure and temperature of the user/seat interface. There are presently seating systems that incorporate rows or a matrix of air cells. An air manifold connected to a battery powered pump alternately shifts the pressure between the cells. The result is a short period of high pressure borne by the skin followed by low pressure. This pressure relief reduces the incidence of pressure sores. Students completing this module may undertake design projects to build an active system.

Case Studies - review of the seating and mobility evaluation, training and follow-up for actual wheelchair users Rationale: Designing for people as a rehabilitation engineer offers challenges that must be learned from experience. Case studies are a great way to compare your ideas for a design with those of a professional rehabilitation team. This team may consist of physical, recreational and occupational therapists, speech pathologists, physiatrists, nurses, rehabilitation engineers, advocates, vocational rehabilitation counselors, social workers, the patient and family. Never overlook the patient's input. They are the most important team member. Case studies are available on videotape.

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