Hearing Aids

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INTRODUCTION

A hearing aid is an electronic device for processing and amplifying sounds to compensate for hearing loss. The primary objective in hearing aid amplification is to make all speech sounds audible, without introducing any distortion or making sounds uncomfortably loud. Modern analog electronics have allowed hearing aids to achieve these goals and provide benefit to many sufferers of hearing loss. However, even if a hearing aid can re-establish audibility of speech sounds, normal speech intelligibility is often not fully restored, particularly when listening in a noisy environment. The development of digital hearing aids has opened the possibility of more sophisticated processing and amplification of sound. Speech processing algorithms in digital hearing aids are being developed to better compensate for the degradation of speech information in the impaired ear and to detect and remove competing background noise.

NORMAL HEARING

The mammalian ear can be divided into three sections: the outer, middle, and inner ears (Fig. 1). The primary function of the outer ear is to funnel sound waves to the tympanic membrane (eardrum) in the middle ear. Vibrations of the eardrum caused by sound pressure waves are transferred to the oval window of the cochlea (inner ear) by the ossicles (articulated bones) of the middle ear. The mechanical system formed by the middle ear helps to create an acoustic impedance match, such that sound waves in the low-impedance air filling the external auditory canal are efficiently transmitted to the high-impedance fluid filling the cochlea, rather than being reflected back out of the ear.

[Fig. 1 about here]

Taking a cross section of the cochlea (Fig. 2) shows that it is divided into three compartments, the scala tympani, the scala media, and the scala vestibule, each of which extends from the base to the apex of the cochlea. Sitting on the basilar membrane (dividing the scala tympani and scala media) is the organ of Corti, which houses the tranducer cells of the cochlea, the outer and inner hair cells. The outer hair cells detect vibrations of the basilar membrane and produce amplification of those vibrations through a mechanism known as electromotility.[2] The inner hair cells synapse onto the auditory nerve fibers which project to the auditory brainstem.

[Fig. 2 about here]

The transduction process from sound waves into neural impulses in the auditory nerve is illustrated in Fig. 3A. Motion of the oval window caused by middle ear movement creates a pressure difference across the basilar membrane. This pressure difference generates a wave that travels along the cochlea from the base towards the apex, causing displacement of the cilia of hair cells. Displacement of the cilia allows ionic currents to flow into a hair cell, modifying the receptor potential (i.e., the hair cell's transmembrane potential). In inner hair cells, this releases neurotransmitter to initiate neural impulses (discharges) in auditory nerve fibers, thus providing information to the brain about the sound vibrations received at the ear.

[Fig. 3 about here]

The mechanics of the basilar membrane are such that high frequency sounds produce traveling waves that peak near the base of the cochlea and low frequency sounds peak near the apex, as illustrated in Fig. 3B. Since each auditory nerve fiber synapses onto only one inner hair cell, each auditory nerve fiber encodes information about the basilar membrane vibration at one position in the cochlea. Consequently, auditory nerve fibers inherit the frequency tuning properties of the cochlea, and each auditory nerve fiber has a best (or characteristic) frequency to which it responds.

Following from the transduction process described above, the auditory periphery is often regarded as a bank of bandpass filters. However, the electromotile behavior of healthy

outer hair cells introduces important nonlinearities into the basilar membrane vibrations and the subsequent representation of sounds by auditory nerve fiber discharges. The predominant nonlinearity is compression, in which the growth of basilar membrane vibrations does not grow linearly with sound intensity but rather is compressed for sound pressure levels between approximately 30 and 90 dB SPL.[5] A related cochlear nonlinearity is suppression, in which the response to a sound frequency component is reduced because of the presence of a more intense component at a nearby frequency.[5] Further nonlinearities are introduced by the inner hair cell and its synapse with an auditory nerve fiber. Auditory nerve responses exhibit adaptation that accentuates the onsets and offsets of sound components, and rectification and saturation are observed in auditory nerve discharge rates. This collection of nonlinearities is important in forming the neural representation of speech sounds.[4,6]

HEARING LOSS

The two most common forms of hearing loss are conductive loss, which refers to dysfunction of the outer or middle ear, and sensorineural loss, in which the inner ear or the auditory pathways of the brain are impaired. The outer and middle ears are largely linear systems, and consequently, conductive dysfunction typically produces a simple linear reduction in the amplitude of acoustic signals as they are transmitted to the inner ear. Thus, loud sounds become quieter and quiet sounds may become inaudible. In contrast, the effects of sensorineural loss are more multifaceted. Since the cochlea is a highly nonlinear system, any impairment of cochlear structures (particularly the inner and

outer hair cells) can lead to substantial distortion in the neural representation of acoustic signals, in addition to loss of audibility. Common causes of sensorineural impairment include exposure to loud sounds, aging, disease, head injury and ototoxic drugs. At present there are no medical or surgical cures for most forms of sensorineural hearing loss.

The loss of audibility of quiet sounds, either from conductive or sensorineural loss, is quantified by hearing thresholds. The hearing threshold is a measure of the sound pressure level required for a tone of a particular frequency to be just audible to a listener. Hearing thresholds are computed using a decibel scale relative to the hearing thresholds of normal hearing listeners, referred to as decibel hearing level (dB HL). A pure-tone audiogram is a plot of hearing thresholds as a function of tone frequency with −10 dB HL at the top and 110 dB HL at the bottom. An example audiogram is given in Fig. 4. By convention, the hearing thresholds for the left ear (LE) are plotted as Xs and those for the right ear (RE) as Os.

[Fig. 4 about here]

The degree of hearing loss is often categorized according to the pure tone average (PTA), which is the average of the hearing thresholds at 500, 1000 and 2000 Hz, the most important frequencies for understanding speech. A typical classification scheme for adults is:

0 to 25 dB HL: normal limits

25 to 40 dB HL: mild loss

40 to 55 dB HL: moderate loss

55 to 70 dB HL: moderate-to-severe loss

70 to 90 dB HL: severe loss

90+ dB HL: profound loss

Alternative classification schemes are also used, particularly for categorizing hearing impairment in children. The example audiogram given in Fig. 4 shows a bilateral, asymmetrical hearing loss, with a mild loss in the left ear (PTA = 33 dB HL) and a moderate-to-severe high-frequency loss in the right ear (PTA = 61 dB HL). The elevation of hearing thresholds with sensorineural loss can be well explained by the loss of auditory nerve sensitivity with dysfunction of the inner and outer hair cells.[4,6]

There are several other aspects of sensorineural hearing impairment that are not captured by the audiogram, including reduction in the dynamic range of hearing, decreased frequency resolution, and decreased temporal resolution. Many of these perceptual impairments can be explained from the direct effects of damaged inner and outer hair

cells on the transduction process in the auditory periphery, but the effects of sensorineural hearing loss on the central auditory pathways of the brain should also be taken into consideration.

The reduction in dynamic range for a hearing impaired listener is often referred to as "loudness recruitment." In this phenomenon, the audibility threshold is elevated at a frequency suffering from hearing loss, but the intensity at which a tone at that frequency becomes uncomfortably loud is relatively normal. That is, there is an abnormally steep growth of loudness with stimulus intensity. This causes difficulties for prescribing gain in a linear amplification scheme for hearing aids: if quiet sounds are amplified enough to restore audibility, then loud sounds may be amplified to uncomfortably-loud levels. It has long been presumed that loudness recruitment is due to more rapid growth of auditory nerve discharge rates with sound level because of loss of cochlear compression. However, recent physiological data contradict this theory and suggest that central mechanisms are required to explain loudness recruitment.[7]

Decreased frequency and temporal resolution and reduced nonlinearities (such as compression and suppression) in an impaired cochlea leads to distortions in the neural representation of a single sound[4,6,8] and to difficulties in separating out different sound sources in the complex acoustic "soundscape" in which we operate daily. For example, it has been shown that after loss of audibility is accounted for, individuals with hearing loss still require a higher signal-to-noise ratio (SNR) to understand speech with the same accuracy as those without hearing loss.[9] Linear amplification schemes cannot

compensate for such distortions; more sophistication speech processing algorithms will be required.

HISTORY OF HEARING AIDS

The first devices for amplifying speech were acoustic ear trumpets, a range of which were available by the 1600s. These used a tapered horn to collect sounds and funnel them to the ear. In addition to providing some amplification, ear trumpets could increase the SNR if the opening of the trumpet was pointed towards the talker. Around 1900, carbon hearing aids were developed that consisted of a carbon microphone, a battery and a magnetic receiver (a miniature loudspeaker), which could produce a gain of around 20 to 30 dB. A pair of carbon hearing aids joined in series, with the coupled magnetic receiver and carbon microphone referred to as a carbon amplifier, could generate even greater gain. In 1920, the carbon amplifier was replaced by a vacuum tube electronic amplifier, which had superior amplification capabilities, but the vacuum tubes required two large batteries to operate. It was not until 1944 that vacuum tubes and batteries became small enough to manufacture a one-piece body-worn hearing aid. During this time, carbon microphones were superseded by magnetic and piezoelectric microphones.

Immediately following the invention of the transistor, head-worn hearing aids using transistor amplifiers instead of vacuum tubes (and consequently smaller batteries) were introduced in the early 1950s. The behind-the-ear (BTE) hearing aid (as shown in the top panel of Fig. 5) was the dominant style for several decades. The invention of integrated

circuit and the electret microphone in the 1960s and 1970s lead to further miniaturization of hearing aid components and the increasing use of in-the-ear (ITE) style hearing aids (shown in the bottom panel of Fig. 5). Continued reductions in component sizes lead to the introduction of smaller in-the-canal (ITC) and completely-in-the-canal (CIC) hearing aids (shown in Fig. 5) in the 1980s and 1990s. In the same time period, steady advances in the electronics of microphones, amplifiers and receivers produced improvements in the fidelity of the audio processing by analog hearing aids. Recently, analog amplifiers have been replaced by digital signal processor (DSP) chips, allowing much more sophisticated amplification and processing of sounds by hearing aids.

[Fig. 5 about here]

MODERN HEARING AIDS STYLES

Current hearing aids are available in a range of styles from BTE to CIC. The typical locations of components in an ITC and a BTE hearing aid are illustrated in Fig. 6. The BTE aid delivers the amplified sound to the ear through an earmold (or earshell) which connects to the BTE ear-hook via a sound tube. Molds or shells are typically custom made from an impression of the hearing aid user's ear, which leads to a comfortable fit in the ear canal and minimization of sound leakage around the mold or shell. Air vents are desirable in an ITE shell or BTE earmold to prevent moisture build-up in the residual canal (i.e., the air between the hearing aid and the ear drum) and to reduce the occlusion effect, in which low-frequency bone-conducted sounds (particularly the hearing aid

user's own speech) resonate in the residual canal. However, venting provides an additional pathway for sounds to enter the ear and for amplified sounds coming from the receiver to exit the ear. This alters the effective frequency response of the hearing aid and provides an additional pathway for acoustic feedback from the receiver to the microphone, which can lead to feedback oscillations (whistling).

[Fig. 6 about here]

The smaller devices are less conspicuous and consequently are more cosmetically appealing to many hearing aid users. However, the larger devices can typically provide superior amplification and sound processing capabilities, because more powerful electronics can be packaged in the hearing aid. Another advantage of the larger styles is the greater physical separation between the receiver sound outlet and the microphone, which may help reduce acoustic feedback oscillations.

LINEAR AMPLIFICATION, COMPRESSION AND FEEDBACK SUPRESSION

The simplest approach to amplification in hearing aids is to apply a linear gain at each frequency that is dependent on the hearing loss at that frequency. As a rule of thumb, a linear gain of 75 % of any conductive hearing loss and around 50 % of any sensorineural loss should be applied. These general rules have been refined greatly through experimental testing to give a number of different amplification prescription formulas, which audiologists use as a starting point when fitting hearing aids. Recent results with a

computational model of the auditory periphery suggest that these fitting rules are effectively restoring the average auditory nerve response to speech.[11]

However, linear amplification may not be optimal, because of distortions that can be produced by waveform peak clipping in the amplifier or receiver at high input sound levels and because of loudness recruitment in an ear with sensorineural impairment. Reduction of the amplifier gain as a function of the input or output signal level via a fast-acting automatic gain control (AGC) is referred to as compression; slow-acting AGC can be used as an automatic volume control. Peak clipping at high levels can be avoided via compression limiting, in which the gain is reduced so that the output level does not exceed some specified limit. Compression that acts over the normal range of speech sound levels is referred to as wide-dynamic-range compression (WDRC). An example input/output curve for a compression system is shown in Fig. 7A, with the gain versus input level shown in panel B. Here, a gain of 30 dB is applied for all input signals that are lower than 40 dB SPL, the knee-point (threshold) of the WDRC. Above 40 dB SPL, the gain of the amplifier is reduced by 10 dB for each 20 dB increase in the input level, corresponding to a compression ratio of 2:1. At an input level of 100 dB SPL, the gain has consequently been reduced to 0 dB, i.e., no amplification is applied. Compression limiting is applied for input levels above 100 dB SPL so that the output level cannot exceed 100 dB SPL.

[Fig. 7 about here]

Compression can be applied identically to all frequencies, referred to as single-channel compression, or individually to different frequency bands, referred to as multi-band compression. The motivation for multi-band compression is that the amount of hearing loss may vary substantially between different frequency bands, and thus the desired compression characteristics may differ between frequency bands. However, multi-band compression will tend to flatten out the frequency spectra of sounds, possibly making it harder to distinguish different sounds. Some multi-band compression schemes have been developed to try to reduce spectral flattening.[12,13]

In addition to the static compression characteristics described by the input/output curve, the rate at which the gain is adjusted with changes in the input (or output) signal level determines the dynamic characteristics of a compression system. The time it takes for a compressor to react to an increase in the signal level is referred to as the attack time, and the release time corresponds to the time taken to react to a decrease in signal level. By adjusting the attack and release times, as well as the static compression characteristics, it is possible for compression systems to avoid distorted and uncomfortably loud signals, to reduce the intensity differences between phonemes or syllables, to provide automatic volume control, to increase sound comfort, to normalize loudness, to maximize intelligibility, or to reduce background noise.[10] Unfortunately, the required compression parameters are very different for each of these goals, and consequently any one compression scheme tends to provide benefit in some but not all aspects of compensating for hearing impairment.

A further use of automatic gain control is to suppress acoustic feedback oscillations, if they can be reliably detected. Reduction of feedback oscillations can also be achieved via automatic phase control, via an internal (i.e., electronic) feedback path to cancel the external (i.e., unintentional acoustic) feedback path, or via frequency shifting by the amplifier to prevent the build-up of oscillations at a particular frequency. With the advent of digital hearing aids, more effective feedback suppression has been obtainable. This has permitted larger vents in earmolds and shells, reducing occlusion effects and providing more comfort to hearing aid users.

ADVANCED PROCESSING ALGORITHMS

To this date, no amplification and compression scheme has been found that can fully compensate for the effects of sensorineural impairment in individuals with moderate to profound hearing loss. Consequently, a number of signal processing algorithms have been developed to make the task of listening easier for hearing aid users. Although some of these processing strategies could be implemented with analog electronics, the processing power and flexibility of DSPs has greatly increased the number and effectiveness of advanced processing algorithms that can be realized in hearing aids.

One class of algorithm now widely used in hearing aids is referred to as single-microphone noise reduction. These algorithms attempt to detect and separate the components of the signal received by the hearing aid microphone that can be attributed to a talker (to whom the listener wishes to attend) and to “background noise” (which the

listener wishes to ignore). Single-microphone noise reduction typically works well when the background noise sources have acoustic properties that are very different from speech and do not vary much over time, but these algorithms have difficulty when the noise sources are speech-like or vary rapidly, as is the case is most real-world listening conditions. A further problem with noise reduction is removing the noise components from a signal without degrading or distorting the target speech components, which could lead to loss of speech intelligibility. Consequently, single-microphone noise reduction algorithms in hearing aids are typically aimed at improving listening ease and comfort, rather than increasing speech intelligibility.

However, noise reduction that leads to intelligibility improvements can be obtained through the use of directional microphones and multiple-microphone algorithms known as beamformers. In this class of algorithms, sounds arriving from a desired direction (typically the direction the hearing aid user is facing) pass through unaltered, but sounds arriving from other directions are attenuated. Some hearing aids utilize microphones with fixed directionality patterns, while others electronically combine signals from multiple microphones such that the directionality can be switched. Adaptive beamformers attempt to detect noise sources and dynamically vary the directionality pattern to cancel them out.

For hearing aid users with very sharply sloping high-frequency losses, transposition of high-frequency components to lower frequencies can provide some improved speech clarity and intelligibility. However, only moderate shifts in frequency are possible, or the

frequency spectra of sounds will become too distorted. An added benefit of frequency transposition is the reduction in acoustic feedback oscillations, as mentioned previously.

In various listening environments, different amplification, compression, noise reduction and directionality schemes may be required. Even before the introduction of fully digital hearing aids, digital memories for analog hearing aid parameters enabled users to manually switch between different parameter sets (referred to as “programs”) in different listening situations. Some hearing aids now use acoustic classification algorithms to detect the listening environment and automatically switch to the preferred program for that situation. The accuracy of algorithms in implementing automatic volume control and feedback suppression has meant that many hearing aids are now available without a manual volume control.

HEARING AID USAGE

Hearing aids can be a viable form of treatment for individuals with any degree of hearing loss, although those with profound losses are good candidates for cochlear implants (see the article *Inner Ear Implants*, pp. 849–855.) However, the audiogram itself is typically not a good guide to how much benefit will be gained from hearing aid use. In some cases, individuals with hearing thresholds small enough to be classified as “normal” may benefit from hearing aids if they experience hearing difficulties in everyday life. The prevalence of hearing loss in most developed countries is reported to be around 10 % of the population. However, regular hearing aid usage is often as low as 20 % in the hearing

impaired population. In addition, many cases of mild to moderate loss may be undiagnosed or not acknowledged by the sufferer. Overall user satisfaction with hearing aids remained around 59 % from 1991 to 2001.[14] While problems such as hearing aid fit, comfort and price contribute to customer dissatisfaction, the major concern is lack of benefit in understanding speech, particularly in noisy listening environments. This indicates that there is substantial room for improvement in amplification and processing of speech by hearing aids.

FUTURE DIRECTIONS IN HEARING AIDS

Since linear amplification and conventional compression schemes do not appear to fully compensate for the effects of sensorineural hearing loss in many hearing aid users, there is growing interest in modeling the effects of cochlear impairment to develop sophisticated nonlinear amplification algorithms that maximally restore the neural representation of sounds to normal.[4,6,8,11,13,15,16] The difficulties in applying this approach include how accurate the model of the ear needs to be, what amplification schemes should be considered, and how impairment of the neural representation should be measured. Simplified cochlear models can lead to more straightforward amplification schemes, but important features of the normal and impaired neural representation of sounds may be neglected. On the other hand, a more detailed neural representation can lead to problems with choosing a suitable amplification scheme and a suitable metric of impairment. The latter difficulty has lead to several investigations into how well models of the ear and subsequent neural processing can predict human speech intelligibility.[17,18]

As we gain a greater understanding of how the ear and brain process speech in normal and impaired cases, we will better know what approaches to amplification can optimally compensate for a particular hearing impairment.

Many hearing aid users have a bilateral loss and consequently wear a hearing aid in each ear. Hearing aid manufactures are now developing wireless data links between digital hearing aids to facilitate coordinated processing by the devices in each ear. At low data transfer rates, this will enable coordinated device programming and synchronized program switching. As higher data rates become feasible, it will be possible to transmit audio signals between hearing aids to perform beamforming and other binaural signal processing based on the signals received by the microphones in each ear. A high-rate wireless data system will also permit transmission of high-quality audio signals to a hearing aid from other audio devices such as telephones, televisions, car and home stereos, public address systems, and MP3 players.

CONCLUSION

Hearing loss can be a debilitating impairment in daily life, greatly affecting social interaction and making many work situations difficult. Until a cure for sensorineural hearing loss is discovered, hearing aids remain the primary treatment option for the hearing impaired population. Continued innovations over more than a century have lead to modern hearing aids that can provide benefit for many hearing impaired individuals. However, hearing aids still cannot fully compensate for the effects of hearing loss to the

same degree that, for example, eye glasses can compensate for the most common forms of visual impairment. Further improvements in amplification and speech processing by hearing aids are under investigation, while at the same time medical treatments for hearing loss are being pursued.

ARTICLES OF FURTHER INTEREST

Hearing Mechanisms, p. 702

Inner Ear Implants, p. 849

Tinnitus Devices, p. 1467

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Hearing Aids

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FIGURE CAPTIONS

Figure 1. Anatomy of the human ear, showing the division of the outer, middle and inner ears. (A) pinna; (B) external auditory canal; (C) tympanic membrane (eardrum); (D) ossicles; (E) Eustachian tube; (F) cochlea; (G) auditory (cochlear) nerve. (Reprinted with permission [PENDING] from Ref. [1].)

Figure 2. A cross section of the human cochlea. (Reprinted from Ref. [3] with permission from Elsevier.)

Figure 3. A) Schematic summarizing the transduction process in the mammalian cochlea. B) Illustration of the tonotopic organization of the cochlea. (Reprinted with permission [PENDING] from Ref. [4].)

Figure 4. An example audiogram showing hearing thresholds for the left ear (LE) and right ear (RE) of an individual with hearing loss.

Figure 5. Modern hearing aid styles: behind-the-ear (BTE); in-the-ear (ITE); in-the-canal (ITC); completely-in-the-canal (CIC). Not shown is the earmold that connects to the BTE

ear-hook via a tube to deliver amplified sounds to the ear canal. (Reprinted with permission from Ref. [10].)

Figure 6. Typical component placement in an in-the-ear (top) and behind-the-ear (bottom) hearing aid. Note that the amplifier is not shown in the ITE diagram. (Reprinted with permission from Ref. [10].)

Figure 7. Example static characteristics of a compression system. A) Input/output curve for a system with wide-dynamic-range compression (WDRC) with a 2:1 compression ratio for input levels above 40 dB SPL and compression limiting at an output level of 100 dB SPL. B) The amplifier gain versus input-level function required to achieve the compression characteristics described in panel A.

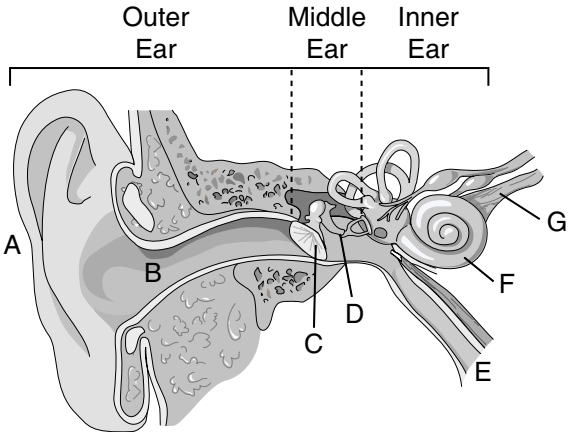


Figure 1

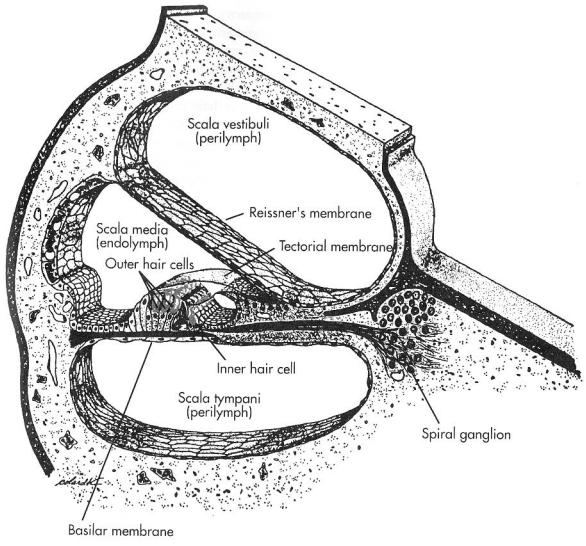

Scala vestibuli
(perilymph)
Reissner's membrane
Scala media
(endolymph)
Tectorial membrane
Outer hair cells
Inner hair cell
Scala tympani
(perilymph)
Spiral ganglion
Basilar membrane

Figure 2

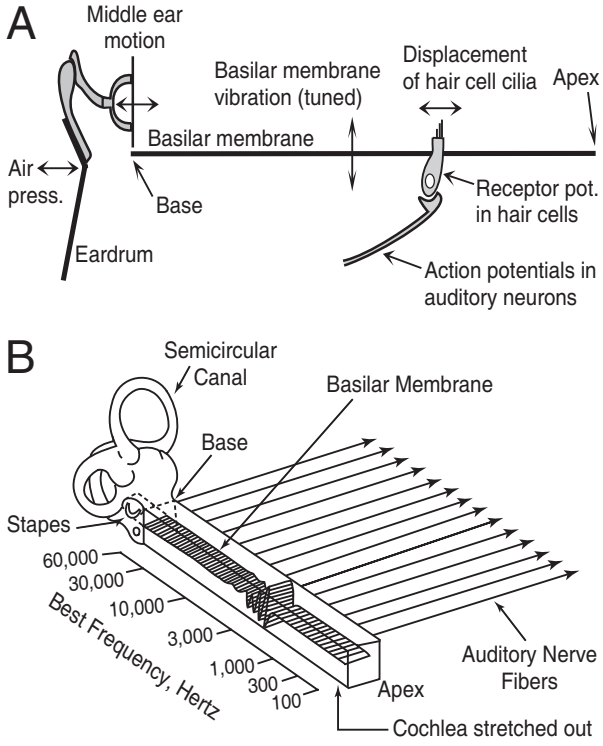

A
Middle ear motion
Basilar membrane vibration (tuned)
Displacement of hair cell cilia
Apex
Basilar membrane
Air press.
Base
Receptor pot. in hair cells
Eardrum
Action potentials in auditory neurons
B
Semicircular Canal
Basilar Membrane
Base
Stapes
60,000
30,000
10,000
3,000
1,000
300
100
Best Frequency, Hertz
Auditory Nerve Fibers
Apex
Cochlea stretched out

Figure 3

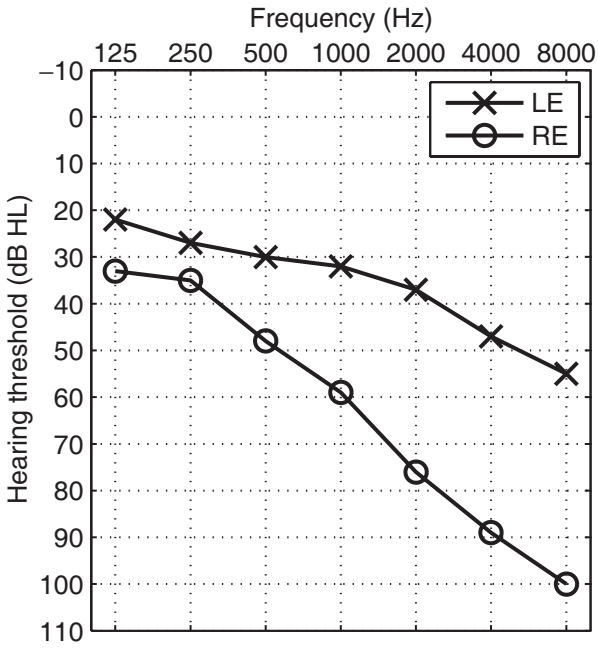

Frequency (Hz)
125
250
500
1000
2000
4000
8000
−10
0
10
20
30
40
50
60
70
80
90
100
110
Hearing threshold (dB HL)
LE
RE

Figure 4

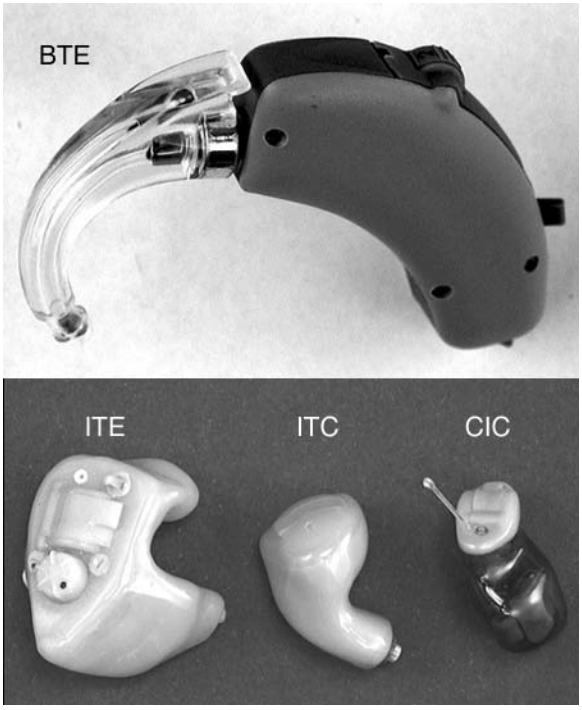


Figure 5

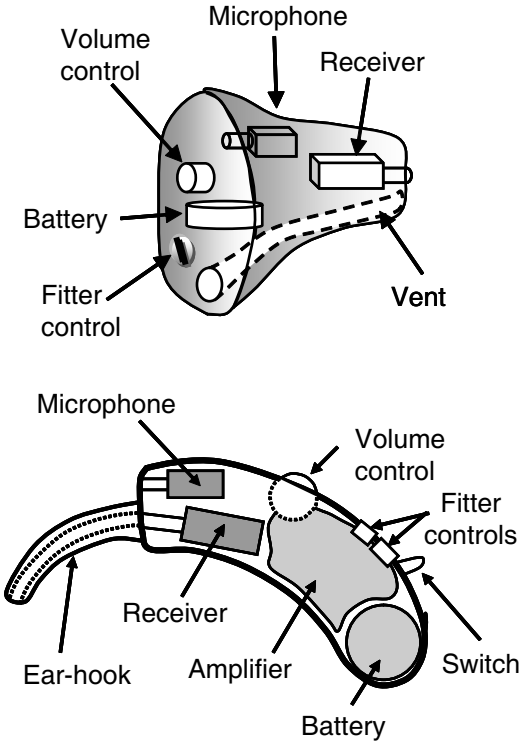

Microphone
Volume control
Receiver
Battery
Fitter control
Vent
Microphone
Volume control
Fitter controls
Receiver
Ear-hook
Amplifier
Switch
Battery

Figure 6

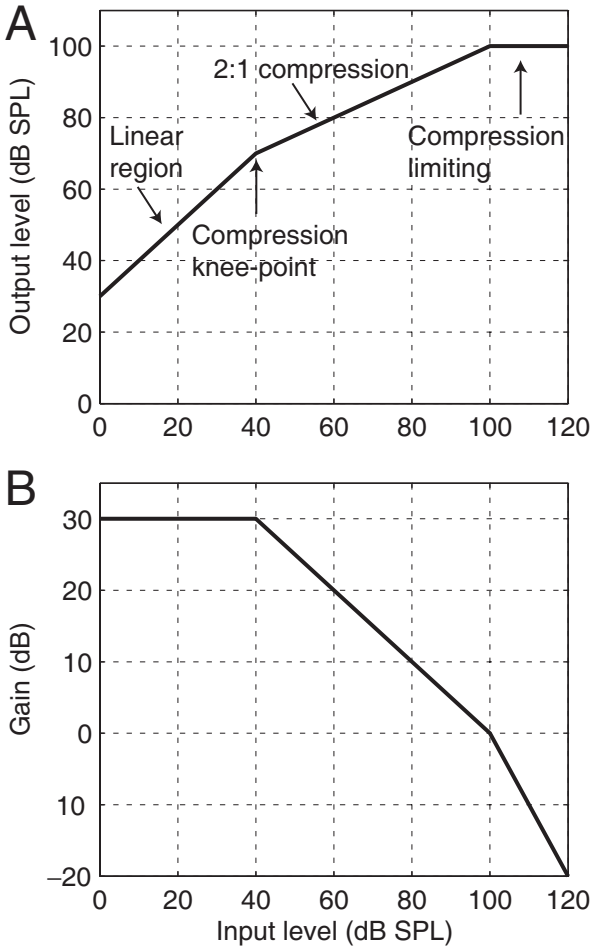

A
100
80
60
40
20
0
Output level (dB SPL)
2:1 compression
Linear
region
Compression
limiting
Compression
knee-point
0
20
40
60
80
100
120
B
30
20
10
0
10
−20
Gain (dB)
0
20
40
60
80
100
120
Input level (dB SPL)

Figure 7