19

Technical Audiology

The enhancement of degraded hearing has been around for a long time. The simplest acoustic device used for hearing enhancement is an ear trumpet. An ear trumpet is a horn that collects sound at its large opening and transmits it to the ear, thus amplifying sound, especially from the direction in which the horn is pointing. When the electronic amplification of sound was invented, it was applied to hearing enhancement. However, it was not until the invention of the transistor and miniaturization of devices that *hearing instruments* became practical. Microprocessors and digital signal processing have introduced new possibilities to enhance the performance of hearing instruments. Modern electronics have also been a prerequisite for the development of *cochlear implants*, in which a major part of the peripheral auditory system is bypassed and sound information is fed electronically to the auditory nerve. It would be reasonable to assume that in the future even more complex means will be developed for the enhancement of degraded hearing.

Electronic technology has also enabled better methods for testing hearing performance. Tuning forks or live speech and whisper sounds may still be used to get a rough estimate of whether the auditory system is functioning in general. However, more advanced methods are needed for more accurate information on hearing performance.

Audiology is a branch of medical science and physiology that studies hearing, balance, and disorders related to them (Martin and Clark, 2006). *Technical audiology* is an interdisciplinary field where the expertise and knowledge of technology and audiology are combined in assessing hearing disorders and enhancing hearing abilities.

This chapter discusses the basic concepts of technical audiology. As background, a brief introduction to hearing impairments and disabilities is given. Then, the methods and technology related to measuring hearing abilities are discussed. After this, hearing instruments (hearing aids and cochlear implants) are discussed.

19.1 Hearing Impairments and Disabilities

Based on an estimation by the World Health Organization (WHO, 2013), 360 million people worldwide have disabling hearing loss, which corresponds approximately to 5% of the

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Ville Pulkki and Matti Karjalainen.

population. WHO defines disabling hearing loss as a hearing threshold shift of 40 dB in the better hearing ear for adults and 30 dB for children. Naturally, an even larger group of people are affected by milder hearing loss. Furthermore, as the average age and the number of the elderly increases, the number of individuals with hearing loss will increase. Thus, a vast number of individuals with hearing impairments, and consequently many different types of hearing impairments, have to be considered in this field.

Impaired hearing can affect the life of the hearing-impaired individual in many ways, depending on the type and degree of the impairment. Most importantly, a hearing impairment can affect orientation in the environment and communication based on sound. Hearing of speech is a fundamental ability for humans, and therefore speech intelligibility has often been used as the basic criterion of the functionality of hearing. However, even a mild hearing impairment affects the perception of details in sound, for example nuances in music. The consequences of a mild hearing impairment will probably be even more important in the future, along with the increased emphasis on the quality of life.

When hearing is substantially impaired, the ability to react to environmental sounds decreases. The inability to react to sonic warning signals poses a significant risk, for example, in traffic. The use of certain machinery requires the user to monitor the sound generated by the equipment in use. In general, sounds have a considerable effect on the perception of experiences and emotions.

The functioning of hearing is of paramount importance in the early years of life, since a hearing impairment can hinder or restrain early language acquisition. A restraint in language acquisition during a certain period early in life can result in a permanent communication disorder. Before the development of proper hearing diagnostic methods, a hearing-impaired child may have been considered mentally challenged, since it was not possible to discover the hearing impairment that hindered learning.

A hearing disability can be temporary or permanent. A temporary hearing disability can be, for example, due to a disease, short-term noise exposure, or excess ear wax in the ear canal. A disease or excessive noise exposure can damage hearing permanently. Permanent hearing disability can also be inborn. For example, the development of senses in the fetal phase is sensitive to any viral infections carried by the mother during pregnancy.

19.1.1 Key Terminology

Hearing loss is the degradation of the sensitivity of hearing, or, in a broad sense, loss of hearing ability in some dimension. Hearing loss is the primary symptom of hearing disorders. A hearing disorder is a structural or functional impairment of the auditory system. The following terms specify the different aspects related to hearing impairments:

- *Disease* – diagnosed and treated medically.
- *Impairment* – diagnosed with audiometry and treated medically.
- *Disability* – is diagnosed via professional evaluation or self-evaluation and treated with rehabilitation.
- *Handicap* – diagnosed with self-evaluation and treated with rehabilitation.

Hearing losses are typically quantified by comparing the hearing thresholds of a patient to hearing thresholds of normal hearing. Normal hearing is denoted by 0 dB in reference to the *hearing level* (that is, 0 dB HL), and the deviations from 0 dB HL are denoted with a hearing threshold shift. A positive hearing threshold shift denotes degraded hearing. The hearing level is the standardized reference that defines the lowest sound pressure level that causes an auditory event at each frequency. Since the sensitivity of hearing varies with frequency, the sound pressure level needed to produce 0 dB HL varies with frequency. The hearing level has been defined by measuring the hearing thresholds for a large number of young individuals with normal hearing. For further definitions, see ISO 389–1 (1998).

Another reference level is the *sensation level* (SL), which defines the hearing threshold for a given individual. For example, if a person has a hearing threshold shift of +10 dB at a given frequency and is presented a pure tone with a level of 30 dB HL, the level of this tone can be expressed as 20 dB SL.

Although hearing loss is typically quantified in terms of hearing threshold shifts, the term ‘hearing loss’ in the broad sense accounts for all the phenomena concerned with the degradation of hearing abilities, which will be discussed in more detail later in this chapter.

19.1.2 Classification of Hearing Impairments

Hearing impairments and disabilities can be classified in various ways. Three different classifications are given below based on social, audiological, and medical criteria.

The social classification of hearing disabilities is based on the overall state of the hearing abilities and on the communication method used by the individual:

1. A person who is *hard-of-hearing* has a hearing loss ranging from mild to severe. Some degree of residual hearing is left, and thus speech communication is usually possible, but a hearing aid may be necessary.
2. A *deaf* person has little or no functional hearing. Sign language is often the primary form of communication.
3. A *deafened* person is a deaf individual who has lost the hearing abilities after learning speech. Lip-reading may provide additional cues in communication.

The classification by the European Working Group on Genetics of Hearing Impairment (1996) defines four groups of impairments based on the average of hearing thresholds measured at frequencies 500, 1000, 2000, and 4000 Hz as follows:

- Mild: 20–40 dB HL
- Moderate: 40–70 dB HL
- Severe: 70–95 dB HL
- Profound: equal to or over 95 dB HL

Based on this classification, the boundary between a normally hearing and hard-of-hearing individual is that the normally hearing person has average hearing thresholds below 20 dB HL. However, different definitions for this boundary exist. For example, in the definition by WHO (2013), a normally hearing person has hearing thresholds lower than 25 dB HL at all frequencies.

The medical classification of hearing impairments is based on the location of the impairment in the auditory system. This classification can be presented, for example, as follows (Martin and Clark, 2006):

1. *Conductive impairments* refer to impairments in the conductive path of the auditory system; that is, the outer ear and the middle ear.
2. *Sensorineural impairments* refer to impairments in the inner ear and in the auditory nerve. Sensorineural impairments can be further divided into cochlear and retrocochlear impairments, based on whether they are located in the inner ear or in the auditory nerve.
3. *Central impairments* refer to impairments in the central auditory nervous system. Their treatment and diagnosis is difficult, since the functioning of the higher levels of the auditory system is less known compared to the peripheral auditory system. Decreased speech intelligibility may be an indicator of a central hearing impairment if the peripheral auditory system has been found to function normally.
4. *Psychic impairments* form a group of hearing impairments where no organic cause can be found.

19.1.3 Causes for Hearing Impairments

A hearing impairment can originate from various causes. A typical outer ear problem is occlusion of the ear canal due to ear wax or a foreign object. Middle ear problems can be due to infections or diseases that damage the auditory ossicles or the eardrum. For example, *otosclerosis* is a disease where new growth of bone stiffens the movement of the auditory ossicles and thus attenuates the sound transmitted to the inner ear. Middle ear damage can also be inborn or due to mechanical trauma, such as head injury.

Sensorineural damage is typically caused by excess noise exposure that damages the hair cells in the cochlea. The hair bundles of the hair cells can be damaged, or the cells themselves can suffer from a metabolic disorder, swell up, or be completely destroyed. Cochlear damage can also be inborn or a consequence of a disease, head trauma, or a tumour in the auditory nerve. Some chemical substances (like kanamycin) are ototoxic; that is, harmful to the inner ear.

Hearing degenerates with age, causing *presbyacusis*, also called *age-related hearing loss*. The sensitivity of hearing is also lower in early childhood compared to adulthood.

In addition, several other factors have been noticed to contribute to the onset of hearing loss (Toppila, 2000). It is important to notice that the joint effect of several factors is often larger than the sum of the factors alone. Such factors include:

- *Vibration*. For example, hand vibration can cause Raynaud's syndrome and weaken peripheral blood circulation. This can affect the blood circulation in the cochlea and thus decrease the tolerance of noise.
- *Smoking*. This does not itself cause hearing loss. However, smoking can have a harmful joint effect with noise, evidently due to the weakening of peripheral blood circulation.
- *Genes*. Sometimes the degeneration of hearing is transmitted in the genes, and hearing loss can occur without any clear reason.

19.2 Symptoms and Consequences of Hearing Impairments

A hearing impairment can result in various symptoms. The main symptom is the degraded sensitivity of hearing (hearing loss), which can be in the form of a hearing threshold shift, decreased discrimination of sound events, and/or distortion of sound. Other symptoms are hyperacusis (oversensitivity to sound) and tinnitus (ringing in the ears). Problems with the sense of balance are related to the vestibular system in the inner ear and are thus part of the field of audiology. However, such problems are not discussed in this book.

19.2.1 Hearing Threshold Shift

A hearing threshold shift can occur due to a conductive or sensorineural hearing impairment. In the case of a conductive impairment, the attenuation of sound is typically linear; that is, high and low sound levels are attenuated by the same amount.

For a sensorineural impairment, the type of attenuation depends on whether the inner or outer hair cells are damaged. If the inner hair cells are damaged, the attenuation is not dependent on the sound level. However, if the outer hair cells are damaged, the resulting attenuation of sound is more prominent at low input levels. This is because when the outer hair cells are damaged, the input-level-dependent amplification provided by them does not function properly. Consequently, loud sounds may be heard normally, but quiet sounds are attenuated. This phenomenon is called *recruitment*. Due to recruitment, a change in SPL translates into an abnormally large change in loudness. Moreover, the range of sound pressure levels from the hearing threshold to the loudness discomfort level is decreased. Consequently, dynamic changes in sound level are overemphasized, and sound is often perceived to be either too loud or too quiet. This is shown in Figure 19.1, which indicates the conceptual loudness-matching functions that visualize the SPL needed for a hearing-impaired person to achieve equal loudness as a normally hearing

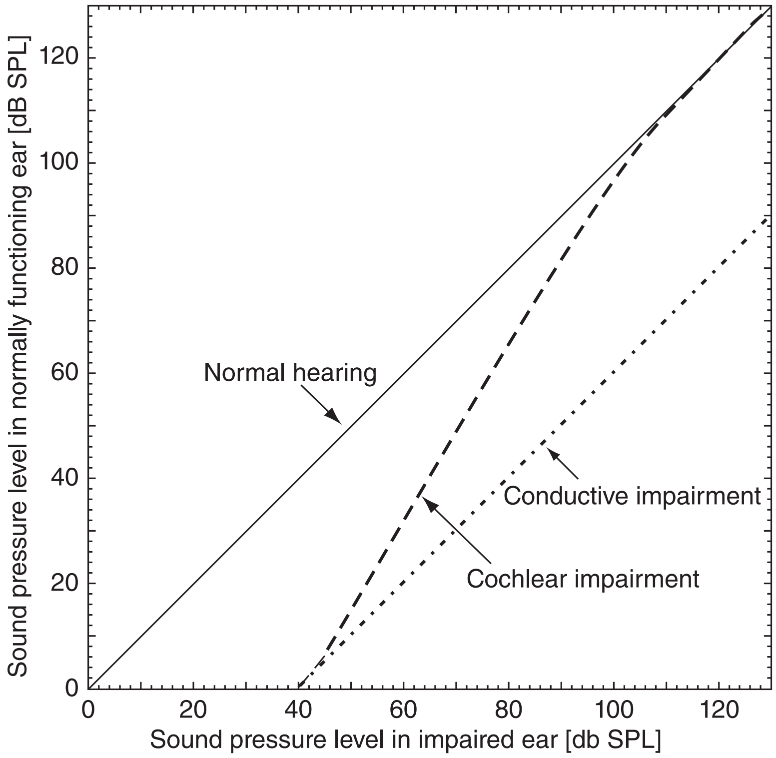

Figure 19.1 Conceptual loudness-matching functions that visualize the SPL needed for a hearing-impaired person to achieve equal loudness as a normally hearing individual: normal hearing, conductive hearing loss, and cochlear hearing loss (recruitment due to outer hair cell damage). Adapted from Moore (2007).

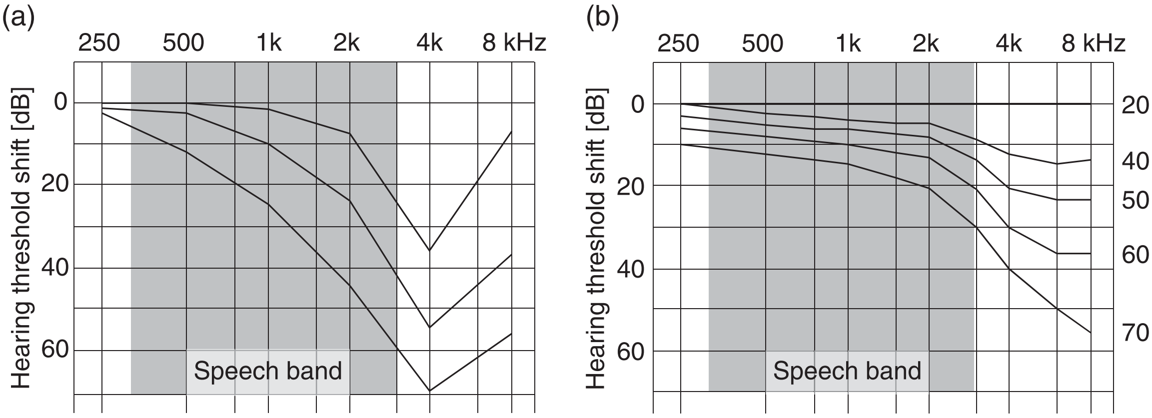

Figure 19.2 Typical hearing threshold shifts induced by (a) excess noise exposure and (b) age. The reference hearing threshold of 0 dB HL denotes normal hearing in young adults.

individual. The curve for patients with cochlear impairment shows a steeper slope, which is a manifestation of recruitment.

A hearing threshold shift, especially when of sensorineural origin, is often frequency dependent. Consequently, hearing can be, for instance, fully functional in one frequency range, have linear attenuation in another frequency range, and recruitment in a third frequency range. This poses certain challenges in the design and individual fitting of hearing instruments, as discussed later in Section 19.5.

Figure 19.2a shows typical hearing thresholds for different degrees of noise-induced hearing impairments. Noise-induced hearing threshold shifts are typically characterized by a notch in the hearing threshold in the 4-kHz range. The effect of age on the degradation of the sensitivity of hearing is individual, but typical hearing threshold shifts due to presbyacusis are shown in Figure 19.2b. The figure shows that age-related hearing loss is typically present at high frequencies. However, the effect of age on hearing loss is difficult to isolate from other factors, such as noise exposure.

19.2.2 Distortion and Decrease in Discrimination

In addition to a hearing threshold shift, a hearing impairment can cause various kinds of distortion. Even at sound pressure levels of normal speech, strong distortion components and echoes may occur, which may decrease speech intelligibility and make listening to music uncomfortable.

Sensorineural hearing impairments often degrade the ability to discriminate sound sources from one another, leading to problems in understanding speech in the presence of background noise or reverberation. The so-called cocktail party effect – the ability to listen to and understand a single speaker in the presence of numerous other speakers – is due primarily to the functioning of the outer hair cells in the cochlea. Thus, if these cells are damaged, the hearing-in-noise ability of the individual is degraded. Problems also arise in reverberant environments even without background noise, since the late reverberation masks the direct sound and early reflections.

Figure 19.3 shows different stages of hair cell damage and the corresponding effect on neural tuning curves. The dashed line in the tuning curve graphs depicts normal hearing. Hearing sensitivity is decreased (that is, the hearing threshold is increased) in a given frequency range

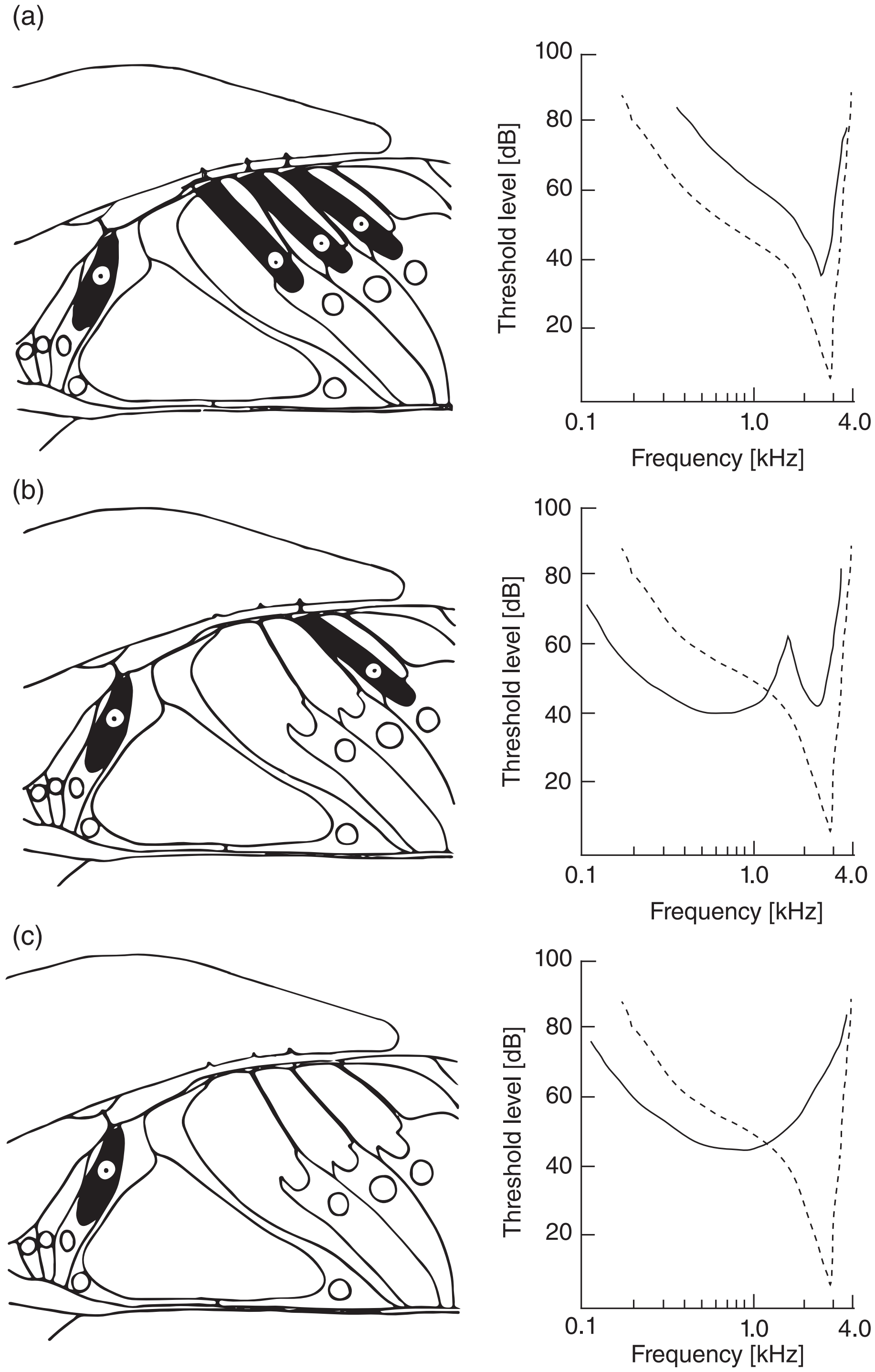

Figure 19.3 Different cases of sensorineural hearing impairments. The hair cell damage in the organ of Corti is depicted on the left. The corresponding neural tuning curve (solid line) compared to normal hearing (dashed line) is depicted on the right. Adapted from Liberman and Dodds (1984).

depending on the degree of hair cell damage in the corresponding place on the basilar membrane. In Figure 19.3a, the inner hair cells are damaged and their sensitivity has decreased, but the outer hair cells function normally. In this case, the threshold level of the neural tuning curves is raised but the frequency selectivity remains normal. A similar tuning curve could result from a conductive hearing impairment. In Figure 19.3b, most of the outer hair cells are damaged, and the corresponding tuning curve has changed shape. In Figure 19.3c, the outer hair cells are completely destroyed, and the corresponding tuning curve is very flat and 'untuned' compared to normal hearing. In this case, the resulting tuning curve depicts the passive amplification characteristics of the basilar membrane without the active effect of the outer hair cells, which, in a normally functioning ear, would increase the frequency selectivity and provide level-dependent amplification, or compression.

The decreased frequency resolution caused by outer hair cell damage effectively makes the critical bands broader with a lower centre frequency and more overlapping. As more energy is added at each critical band, each band receives input from a wider frequency range, the frequency masking effect is stronger. The result is degraded speech intelligibility in noise.

19.2.3 Speech Communication Problems

In addition to the effects of sensorineural impairments discussed above, speech discrimination can also be degraded due to impairments in the processes specifically responsible for speech processing. An individual may hear many sounds without problems but may still be unable to understand speech. For example, there may be problems in analysing the temporal structure of speech. This can depend on the pace of the speech, so that slow speech is intelligible, but as the pace increases, the auditory system cannot structure the speech into linguistic units.

Hearing impairments can also result in problems in speech production due to lack of feedback of the produced sound (see Section 17.8.3 for a review of the effect of feedback on speech production). Moreover, the ability to speak may be entirely restrained due to inborn deafness.

Speech as a form of linguistic communication is a complex phenomenon, which makes measuring the level of speech communication abilities difficult. Since speech is the most important method of communication for humans, speech audiometric methods have been developed to measure these abilities.

19.2.4 Tinnitus

Tinnitus, a ringing in the ears, is a common phenomenon that occurs in most people at some point in their lives, at least for a short while. Tinnitus is an auditory event that is not related to any external sound event. Tinnitus can present itself in various forms of perceived sound. A typical tinnitus sound is a steady pure-tone-like sound, but it can also be, for example, broadband or band-limited noise. The sound may also vary in time, for example, along with heartbeats, in which case the tinnitus is probably due to blood pressure changes in the ear. Another typical cause for tinnitus is excess noise exposure.

Several theories have been put forward explaining the origin of tinnitus in the auditory system (Jastreboff, 1990). Tinnitus may originate from the malfunction of some neural process in the auditory system with no actual vibration. It may also originate in the cochlea, even to such an extent that the basilar membrane vibrates and the tinnitus sound can be measured in the ear canal. In this case, the functioning of the outer hair cells in the cochlea has become unstable and thus a constant oscillation is present even without an acoustic input.

Loud tinnitus is a psychologically difficult impairment and can have a significant effect on the quality of life for the individual. There is no known cure for tinnitus. However, methods exist for treating tinnitus. One approach is to use a tinnitus masker to generate a sound in the ears that makes the tinnitus sound less annoying. Another approach is to use a combination of counselling and sound therapy to habituate the individuals to the tinnitus sound in order to reduce their awareness of it (Jastreboff *et al.*, 1996).

19.3 The Effect of Noise on Hearing

The effect of noise on hearing and the prevention of noise-induced hearing loss are important research subjects in the fields of acoustics and audiology. Noise is the most common cause for work-related impairments (Toppila, 2000). Avoiding exposure to noise is difficult in modern society. Excess noise exposure can cause a hearing impairment, but even lesser exposures can cause mental stress and consequent symptoms.

19.3.1 Noise

Noise is harmful or disturbing sound. This definition embodies two aspects: harmfulness and disturbance. Harmfulness is the more objective of the two: noise can be harmful even if it is perceived as comfortable and non-disturbing. The effects of harmful noise, for example hearing loss, can be measured using audiometric methods. The disturbance caused by sound and especially annoyance are more subjective concepts and can be measured, for example, in terms of a decrease in work efficiency or the ability to concentrate. As already discussed in Section 17.10, *disturbance* can be defined by how much a sound disturbs some action, while *annoyance* is a more subjective concept, which can be estimated based on individual experience. Subjective handicaps may have further indirect consequences, such as mental and physical illnesses.

Good examples of conflicts between subjective and objective points of view of noise are loud music, loud motorsports, shooting, and other noisy hobbies. These can be considered positive experiences even when the risk of hearing loss is obvious.

An essential question in the field of noise control is how to estimate the negative effects of noise. The majority of the related research discusses the risk of hearing loss, while the disturbance and annoyance caused by noise are less studied. There are two reasons for this. First, the risk of hearing loss is primarily a physiological phenomenon, where the induced risk is mainly predicted by the properties of the sound. Second, disturbance and annoyance are more dependent on the individual and on the situation. For example, loud music may be a ‘fantastic aesthetic experience’ or ‘terrible noise’ depending on the individual and on the situation.

The primary factor determining the negative effects of noise is *noise exposure*, which depends on the level and on the exposure time. When measuring noise exposure, the frequency-dependent sensitivity of hearing should be taken into account. In practice, this is typically done by measuring the frequency-weighted sound pressure level. Figure 9.3 on page 156 shows the A-, B-, C-, and D-weightings, which approximate the inverted shape of the equal loudness curves at different sound pressure levels (see Figure 9.2). The A-weighting has become the most widely used frequency-weighting, although it is not optimal for all sound pressure levels. However, the A-weighting seems to describe the risk of hearing loss quite well.

Noise exposure can be measured using the *equivalent sound pressure level*, which takes into account the temporal variation of the sound pressure level. The equivalent sound pressure level L_{eq} is the root-mean-square level averaged over a certain time period:

$$L_{\mathrm{eq}} = 10 \log_{10} \frac{\sum \Delta t_i \, 10^{L_i/10}}{T}, \qquad (19.1)$$

where t_i are the time periods with sound pressure levels of L_i [dB SPL] and T is the total time with which the sound exposure is normalized to the equivalent sound pressure level. T is usually 8 hours, corresponding to a normal working day. An integrated sound pressure level measure L_{eq} can be computed so that the sum becomes an integral when Δt_i tends to zero.

Research has shown that noise with considerable impulsive content has a greater risk of causing hearing loss compared to steady noise. Therefore, it is reasonable that in the case of impulsive noise, this effect is taken into account. A simple rule is that when impulsiveness exceeds a certain threshold, an extra 10 dB is added to the equivalent sound pressure level.

19.3.2 Formation of Noise-Induced Hearing Loss

The threshold of pain in hearing is around 130 dB SPL. Sounds louder than this can cause permanent damage to hearing even over short durations. For example, a starting pistol produces a peak sound pressure level of approximately 140 dB SPL, and more hefty weapons produce even louder sounds. A single strong peak in sound pressure can tear the eardrum, break the auditory ossicles, or destroy the structures of the inner ear immediately. An impulse even at a lower level can cause permanent tinnitus.

Still, typical noise-induced hearing loss develops over the course of time. The loss can remain unnoticed until speech intelligibility begins to suffer. For long-term noise exposures, implying years or decades of exposure, the daily risk threshold has often been quoted to be 85 dB SPL (A-weighted, 8 hours a day), with the criterion of impaired hearing being a hearing threshold shift of 25 dB in middle–high frequencies. This kind of threshold shift already may decrease speech intelligibility. If healthy hearing is to be assured even after years of exposure to noise, the A-weighted risk threshold should be decreased to 80 dB, or even 75 dB. This is because the risk of hearing loss is somewhat individual, and, for example, music listening may suffer even with smaller hearing impairments.

19.3.3 Temporary Threshold Shift

As discussed earlier, a high enough noise exposure can cause the sensitivity of hearing to decrease, and consequently the hearing thresholds to increase. If the exposure is under certain limits, the hearing thresholds can recover after the exposure, fully or to some extent. This phenomenon is called the *temporary threshold shift* (TTS). The level and duration of the TTS are affected by the level and duration of sound. Figure 19.4a shows the increase in hearing threshold as a function of time for different sound levels. Even a sound of 70 dB SPL is capable of generating a minor TTS when the exposure time is several hours. Sound pressure levels of over 100 dB SPL increase the hearing threshold rather quickly.

When the noise exposure has ended, the hearing threshold begins to recover. Figure 19.4b shows the decrease in the hearing thresholds after a TTS caused by a long exposure to sound at different SPLs. The recovery time can be from minutes to several days. As a general rule, one should allow about double the exposure time for recovery, or a time when hearing has recovered to normal. A sufficiently large exposure to noise leads to a permanent hearing threshold shift and the recovery to normal hearing does not happen.

Although hearing recovers to normal after a TTS, it is a sign of a risk of hearing loss, especially in the case of repeated temporary threshold shifts or a single significant TTS. Thus a good rule of thumb is that, for example after a rock concert, hearing should be normal by the next morning at the latest. In addition to TTS, tinnitus or sound distortion effects after an exposure to noise are signs of overload in the auditory system.

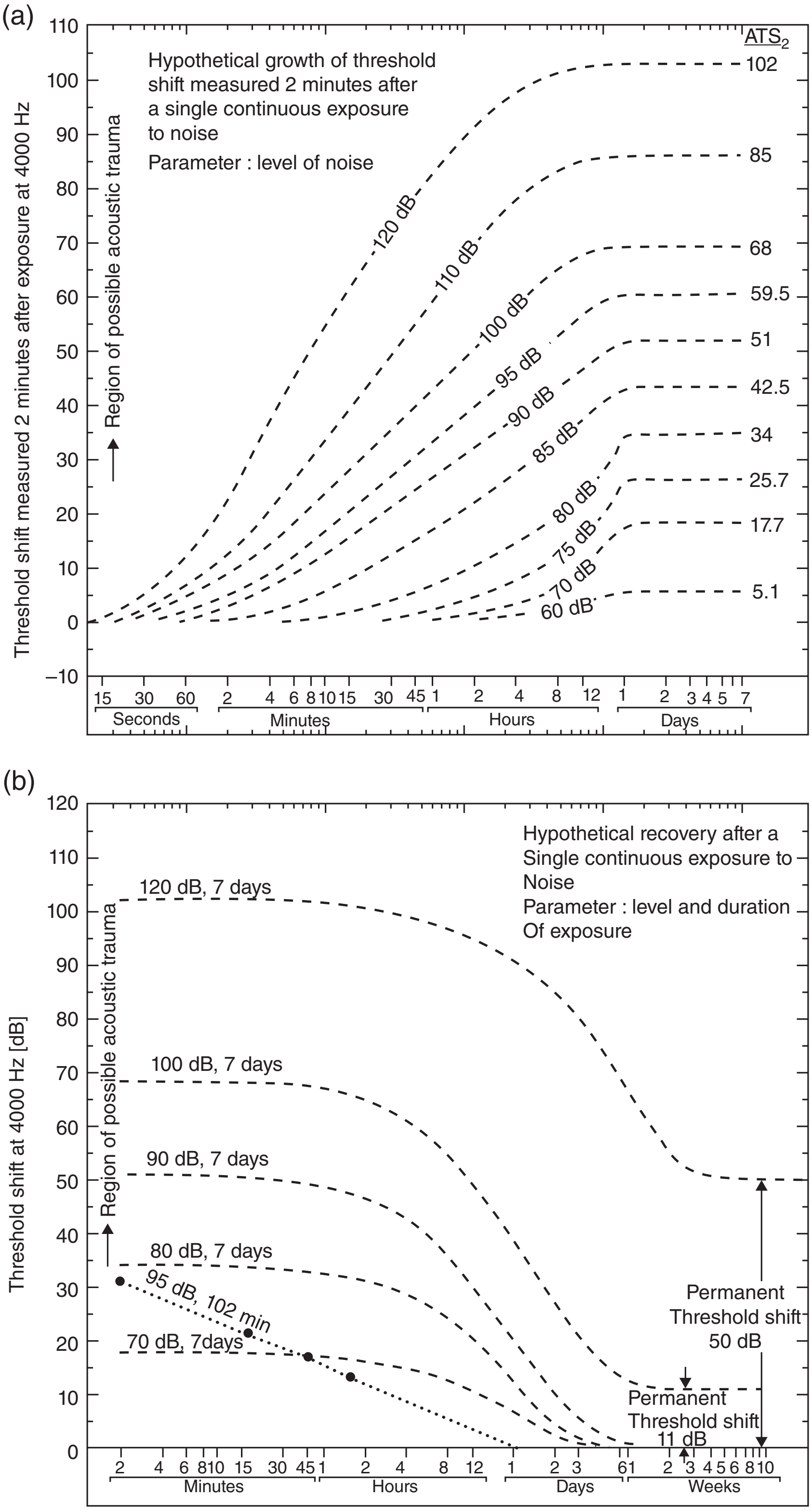

Figure 19.4 (a) The hypothetical formation of a temporary threshold shift (TTS) at different exposure levels as a function of exposure time. (b) The hypothetical recovery from TTS as a function of time, after a long-term noise exposure at different levels. Adapted from Miller (1974), and reprinted with permission from The Acoustical Society of America.

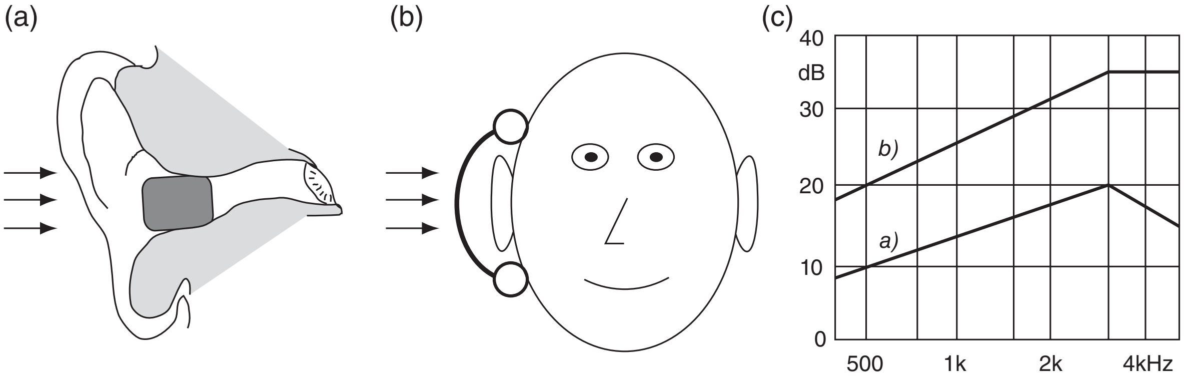

Figure 19.5 (a) Earplug-type of hearing protector, (b) earmuff-type of hearing protector, and (c) typical attenuation characteristics for earplugs and earmuffs. Adapted from Toivanen (1976).

19.3.4 Hearing Protection

When controlling noise that is potentially dangerous to hearing, countermeasures should be taken in the following order of priority:

1. *Decrease the noise emission of the noise source*. For example, design a quieter machine.
2. *Attenuate the transmission path*. For example, encapsulate the machine that produces noise.
3. *Personal hearing protection*. Use when the procedures above are not sufficient.

Hearing protectors attenuate sound that reaches the eardrum. They can be divided into different groups depending on their type:

- *Earplug*: a plug that blocks the ear canal entrance (Figure 19.5a). It is typically made of plastic foam or wax. Individually moulded earplugs exist too. For effective attenuation, the plug should be rigid enough and seal the ear canal properly.
- *Earmuff*: constructed from a stiff cup that is tightly placed against the head with padding so that the whole outer ear is covered (Figure 19.5b). These types of hearing protectors are the most common in situations where significant attenuation is needed. The increase in mass and internal volume of the earmuff increase the attenuation achieved with the protector.
- *Acoustic helmet*: a helmet that covers the whole head or part of the head and can provide extreme attenuation of sound.
- *Active hearing protection*: can be used with any of the above-mentioned passive types of protectors. Active hearing protectors are equipped with an electroacoustic system that can be used in various ways. One method is to reproduce the incoming sound with opposite phase from the earphone, so that this sound cancels out some of the sound coming through the ear protector enclosure. This enables the attenuation of a passive protector to be increased. With well-designed active earplugs, efficient attenuation can also be achieved in the low frequencies. Another approach is to attach active hearing protection to heavy earmuffs and provide amplification only to quiet sounds. In this way, for example, speech communication can be possible while using earmuffs, but loud sounds (e.g., sudden impulsive sounds) are attenuated passively.

Figure 19.5c shows typical attenuation characteristics of earplugs and earmuffs. However, different models vary considerably in their attenuation characteristics. A good, well-fitting earplug may provide better attenuation than a light earmuff. By combining the use of earplugs and earmuffs, even more attenuation can be achieved.

It is hard to achieve good attenuation in the low frequencies unless the protector is very well sealed and of appropriate construction. Thus, the correct use of the hearing protectors is also important. Otherwise, the attenuation performance may be significantly worsened. An earplug should be fitted deep enough and so that it seals the ear canal properly. When using earmuffs, one must ensure that there is no gap between the earmuff padding and the head, for instance, due to hair or glasses. With an improper fit, an earmuff can even form a Helmholtz resonator, which amplifies sound at certain frequencies.

Hearing protection should be used constantly when exposed to loud noise, because even a short time without hearing protection can cause significant noise exposure and thus a considerable risk of hearing loss.

19.4 Audiometry

Audiometry is the science of measuring the functioning of the auditory system. The first measurement methods were based on testing the patient's response to a sound produced, for example, by a tuning fork or by live speech. Nowadays, several dedicated methods and pieces of equipment are used to achieve reliable and repeatable results. The term *audiometer* refers to a pieces of equipment used to conduct audiometry.

The techniques to estimate the functioning of the auditory system can be divided into two categories. Tests where the subject has to listen actively to the sound events and to co-operate in responding to them can be said to be subjective methods. These tests are commonly used, and they can be conducted easily with most people. Some cases require objective methods, where responses do not depend on whether the test subject is listening or not. Such tests can be utilized with infants or with people who are unable or unwilling to co-operate. Furthermore, objective tests can be useful in monitoring the state of the auditory system during, say, a surgical operation.

19.4.1 Pure-Tone Audiometry

Pure-tone audiometry (ISO 8253-1, 2010), as the name suggests, uses pure tones or other narrowband stimuli as test signals to measure the frequency-specific hearing thresholds. A *pure-tone audiometer* is a device that varies, either manually or automatically, the level of the test signal in an adaptive procedure seeking the hearing threshold. Pure-tone audiometry is usually performed by presenting the test stimuli via calibrated headphones, to which the patient responds orally or by pressing a button. This procedure is done separately for both ears, typically at least at the frequencies 250, 500, 1000, 2000, and 4000 Hz, and possibly also at 125 Hz and 6000 Hz and/or 8000 Hz. Békésy audiometry uses a slow frequency sweep, thus measuring a continuous hearing threshold curve as a function of frequency. The steepness of the hearing threshold curve can be obtained from the results of Békésy audiometry.

Pure-tone audiometers are manufactured for different purposes and with different precision requirements. In hearing screening, for example, the aim is only to discover possible significant deviations from normal hearing. Dedicated screening audiometers are available for this purpose. Clinical audiometers (Figure 19.6) are used for more precise measurements. Clinical and diagnostic audiometers can include many special functions. Simple audiometers for home use (for example, computer software) also exist. For reliable results, it should be ensured that the masking effect of background noise does not increase the measured hearing thresholds,

Figure 19.6 An audiometer with audiometric headphones and a patient-response button. Courtesy of Teemu Koski.

and that the transducers are properly calibrated. The hearing threshold curve as a function of frequency measured in pure-tone audiometry is called an *audiogram*.

19.4.2 Bone-Conduction Audiometry

Sound is transmitted to the inner ear not only through the outer and middle ear (that is, air conduction), but also through the bones of the head (that is, via bone conduction). For sound that propagates in air, bone-conducted sound is quieter than air-conducted sound. Thus, air-conducted sound generally dominates in the perception of sound. However, vibration that is coupled mechanically to the bones of the head is effectively transmitted to the inner ear via bone conduction. Consequently, bone conduction has a significant effect, for example, on how one's own voice is perceived.

The bone-conduction phenomenon is utilized in bone-conduction audiometry. In contrast to air-conduction pure-tone audiometry (discussed above), the test stimuli are fed as vibration to the skull at the back of the ear in bone-conduction pure-tone audiometry. The bone-conduction hearing thresholds are then measured in a similar manner to air-conduction pure-tone audiometry. If the bone-conduction threshold is normal but the air-conduction threshold is elevated, the hearing impairment is likely to be conductive. If the bone-conduction threshold has increased by the same amount as the air-conduction threshold, the impairment is likely to be sensorineural or due to a problem at some higher level of the auditory system.

19.4.3 Speech Audiometry

Speech is the most important form of sound-based communication. Degraded speech intelligibility is a common practical consequence of many kinds of hearing impairment. In some cases, these impairments can be diagnosed using pure-tone audiometry. However, speech intelligibility may be decreased even when hearing thresholds are normal and no organic impairment is found. *Speech audiometry* methods (ISO 8253-3, 1996) have been developed to measure speech intelligibility directly.

In speech audiometry, the patient tries to identify words, sentences, or other speech sounds he or she is presented with. Speech can be reproduced to the patient over headphones or loudspeakers. Typically, the target signal level is either constant or varying to measure the percent-correct score or the detection threshold, respectively. Moreover, the speech pace can be varied, because in some cases the individual is able to understand only slow speech. Dedicated speech corpora have been developed for speech audiometry for many languages, for example Kollmeier and Wesselkamp (1997) and Nilsson *et al.* (1994).

Speech audiometry can also be conducted with background noise. These speech-intelligibility-in-noise-measurements aim to assess problems that the hearing impaired commonly have in communication situations involving background noise.

19.4.4 Sound-Field Audiometry

In *sound-field audiometry* (SFA), loudspeakers are used instead of headphones for the reproduction of the test stimuli. Compared to headphone-conducted audiometry, SFA demands more from the equipment and facilities, but in turn enables test conditions that are not possible or practical with headphones. The test stimuli can be narrowband, as in conventional pure-tone audiometry, or, for example, speech in silence or with background noise. The standard ISO 8253-2 (2009) defines the use of narrowband stimuli in SFA.

The primary motivation for using SFA is the limitations in headphone-conducted audiometry. The first limitation, for a certain group of test subjects, is the acoustic coupling between the sound source and the test subject. Depending on the type of hearing instrument and the microphone placement, it is often difficult to achieve a constant and controlled acoustic coupling between the audiometric headphones and the hearing instrument microphone. Small children might not tolerate the use of headphones, again causing uncontrolled acoustic coupling. The second limitation is the realism of the audiometric test. Testing in a sound field, in contrast to the use of headphones, takes into account the spatial attributes of sound and hearing and allows test conditions to more closely resemble real-life situations relevant to the patient.

The simplest SFA set-up consists of one or two loudspeakers. More complex SFA systems aiming to reproduce real or realistic sound scenes have been suggested, for example, by (Favrot and Buchholz, 2010; Koski *et al.*, 2013; Seeber *et al.*, 2010). In addition to audiometry, SFA methods are used to evaluate and compare the performance of hearing instruments (see, for example, (Minnaar *et al.*, 2013)).

19.4.5 Tympanometry

Tympanometry (Campbell and Mullin, 2012; Martin and Clark, 2006) is an objective way to measure the status of the eardrum and the middle ear. It measures the acoustic impedance of the eardrum – the tympanic membrane – to assess the mobility and the pressure difference across the eardrum.

Figure 19.7 shows the measurement set-up for tympanometry. A test sound is presented to the ear canal with an earphone, and the sound pressure in the ear canal is measured with a microphone. The changes in the impedance of the eardrum are reflected in the measured sound pressure. In other words, when much of the sound is reflected back from the eardrum,

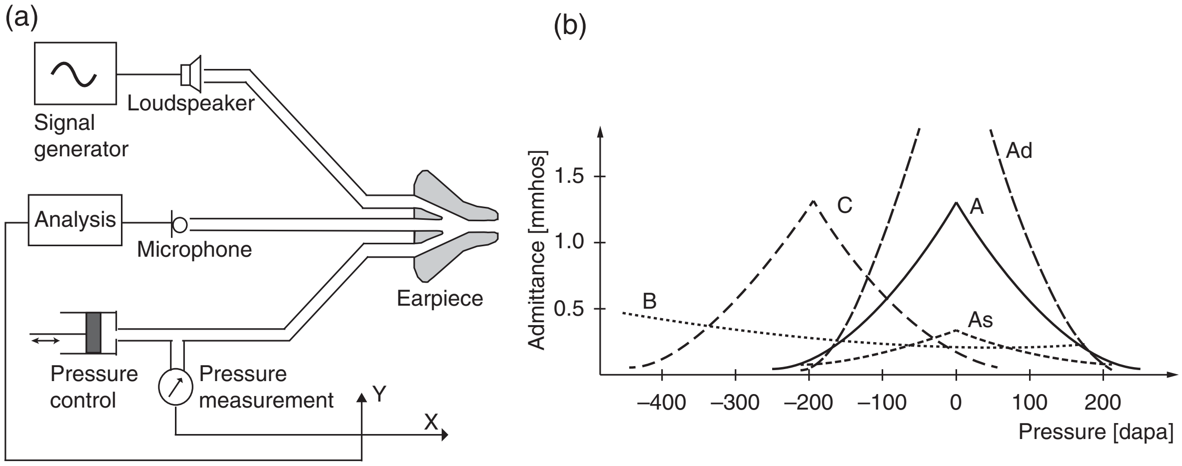

Figure 19.7 (a) The equipment set-up in tympanometry and (b) a conceptual tympanogram, showing examples of different responses to tympanometry measurement. A = normal middle ear function, As (for A-shallow) = stiffened middle ear system, Ad (for A-deep) = flaccid eardrum, B = fluid in the middle ear or perforation of the eardrum, C = negative pressure in the middle ear. Data adapted from Campbell and Mullin (2012).

the acoustic impedance is high, indicating a stiff eardrum (low compliance), and when little sound is reflected back, much of the sound is transmitted to the middle ear, and the acoustic impedance is low, indicating a more mobile eardrum (high compliance). During the playback of the test sound, the external static air pressure is swept over a range of negative and positive pressures and the value of compliance is measured as a function of the static air pressure.

The external pressure where the maximum compliance is found corresponds to the pressure in the middle ear. If this pressure differs considerably from the atmospheric pressure, there is a pressure difference across the eardrum. The absence of a compliance peak may imply fluid in the middle ear. The level of the maximum compliance indicates the condition of the eardrum and the auditory ossicles. For example, a low maximum compliance may indicate stiffness in the auditory ossicles, while a high maximum compliance may indicate a flaccid eardrum.

Even a relatively small pressure difference across the eardrum attenuates the sound transmitted to the inner ear, especially at the low frequencies. A large pressure difference can be very painful. A pressure difference may originate, for example, from an infection or static changes in the air pressure, for example during a flight. The Eustachian tube, which connects the middle ear to the pharynx, normally equalizes the air pressure between the middle ear and the outer ear. However, the tube may be narrow in some individuals, or it may be blocked due to infection, and thus the pressure may not be equalized.

19.4.6 Otoacoustic Emissions

Otoacoustic emissions, discussed in Section 7.5, can be used to objectively evaluate the functioning of the auditory system (Martin and Clark, 2006). An *evoked otoacoustic emission* (EOAE) should be observable if the auditory system is functioning normally; missing EOAEs indicate a conductive or cochlear defect. Such an indication does not specify whether the defect is due to the cochlea not responding to the stimuli or due to attenuation in the conductive path. EOAEs present in cases of sensorineural hearing loss suggest retrocochlear problems.

Transient-evoked otoacoustic emissions (TEOAEs) are measured using brief acoustic stimuli that stimulate a wide area of the cochlea, resulting in a broadband response for a normally functioning auditory system. *Distortion-product otoacoustic emissions* (DPOAEs), in turn, are measured with two tones that differ in frequency, for which a normally functioning cochlea responds with an emission in additional frequencies.

19.4.7 Neural Responses

The functioning of the auditory system can also be assessed by measuring neural responses from different parts of the auditory path. *Electroencephalography* (EEG) and *magnetoencephalography* (MEG) are non-invasive methods to measure such responses. In EEG, the weak electrical potentials produced by brain activity are measured with electrodes that are placed at different locations on the scalp. MEG measures the weak magnetic fields produced by brain activity. Typically, sound stimuli are presented to a subject, and the EEG or MEG responses evoked by the stimuli are measured.

19.5 Hearing Aids

A *hearing aid* (Dillon, 2012) is basically a miniature sound reproduction system with signal processing. Its purpose is to amplify and process sound for the user to compensate for the effects of a hearing impairment. The eventual aim is to enhance the hearing and communication performance of the user. A challenge in hearing-aid design is to achieve a suitable type of amplification that takes into account individual needs in different listening scenarios. Ensuring an adequate SNR for the user is essential. Other design objectives are, for example, comfort and ease of use.

The simplest possible hearing aid is a microphone placed near the ear, a linear amplifier with constant gain over the entire audible frequency range, and a miniature loudspeaker providing the amplified sound to the ear canal. This may suffice to compensate for a conductive hearing loss similar at all frequencies, but not for cases involving frequency- and level-dependent attenuation. Thus, modern hearing aids include frequency-dependent amplification, compression, and other signal processing features.

19.5.1 Types of Hearing Aids

Many types of hearing aids are available, varying in their operating principle and structure. The main types are listed below:

- *Behind-the-ear (BTE) hearing aid*, Figure 19.8a: The instrument enclosure is located behind the ear. BTE aids transmit sound to the ear canal via a tube. The receiver-in-the-canal (RITE) aid is a variation on the BTE, where an earphone is placed in the ear canal and the sound is transmitted there electrically, instead of through a tube as in BTE devices.
- *In-the-ear (ITE) hearing aid*, Figure 19.8b: The instrument is fitted in the concha.
- *In-the-canal (ITC) hearing aid*: The instrument is fitted in the ear canal. A variation on the ITC hearing aid is the *completely-in-the-canal (CIC) hearing aid*, which fits completely in the ear canal (Figure 19.8c).

More hearing aid types exist in addition to the ones listed above. Many of them are variations on the BTE, ITE, and ITC structures, but other approaches are also used. For example, in an *eyeglass instrument*, the aid is embedded into spectacles. This allows, for example, a microphone array to be placed along the frames, thus enabling highly directional sound-capturing

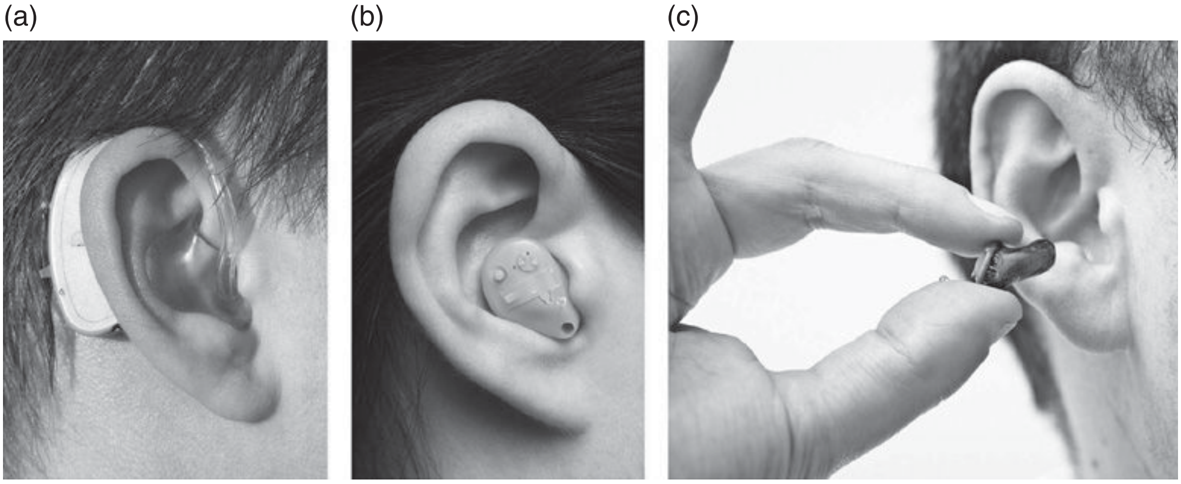

Figure 19.8 Types of hearing aids: (a) behind-the-ear (BTE) hearing aid, (b) in-the-ear (ITE) hearing aid, and (c) completely-in-the-canal (CIC) hearing aid

properties. The main unit of the hearing aid can also be completely detached from the head of the user. This kind of *body-worn instrument* was used especially in the past, when it was not possible to fit all the necessary equipment in a small casing.

19.5.2 Signal Processing in Hearing Aids

Modern hearing aids utilize various signal processing features, aimed at matching the device with the individual needs of the user (Kates, 1998). Digital signal processing enables many features that could not be achieved with analogue technology. Although full coverage of these techniques is beyond the scope of this book, a general overview of the most relevant techniques is given below.

The degree and properties of a hearing impairment can vary considerably for different frequencies. Thus, hearing aids often employ multi-band processing, where the input signal is divided into several frequency bands using a filter bank and the amplification and processing are done separately in the bands. This allows all the parameters related to amplification and other processing to be optimized for each frequency band, based on the measured hearing performance.

Communication situations involving background noise or reverberation often cause problems for the hearing impaired. An individual with impaired hearing might struggle in these kinds of situations even though speech intelligibility in quiet situations is normal. The hearing impaired often need a higher SNR to achieve the same speech intelligibility as normally hearing individuals. Plain amplification amplifies both the desired signal and the unwanted noise, and therefore it does not solve the problem of degraded speech intelligibility in the presence of background noise. Thus, several approaches have been developed to increase the SNR in the hearing aid output and so enhance the intelligibility in acoustically complex environments.

Gain control

Linear amplification increases the output level in the same proportion to the input level. If the input level increases by, say, 1 dB, the output is increased by 1 dB as well. This poses a problem for high input levels: either the level exceeds the dynamic range of the device or the

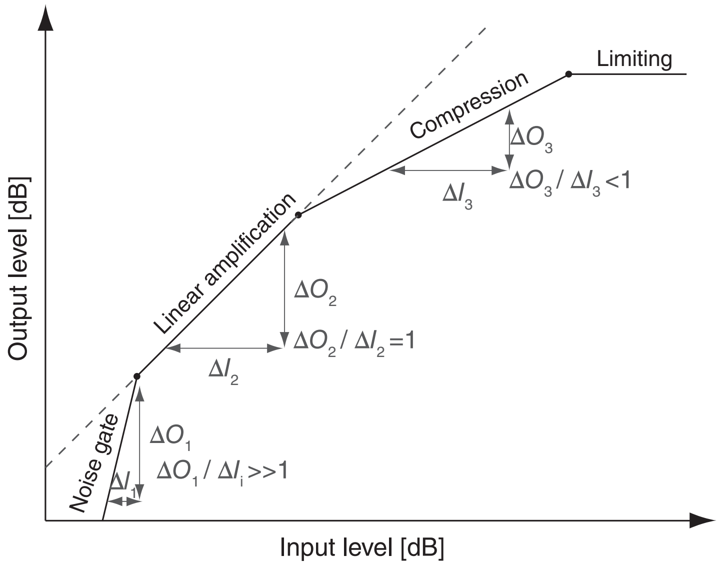

Figure 19.9 An amplification curve with high-level compression, linear amplification, and limiting. The changes in the level of input and output in dB are denoted as ΔI and ΔO, respectively. Courtesy of Teemu Koski.

amplified output can be so loud as to cause a risk of further hearing loss. Thus, the amplifiers in hearing aids are equipped with some kind of *limiter* circuit that limits the output amplitude of the amplifier to a certain value. Unfortunately, a sharp limiter causes the signal to distort as the level exceeds the pre-set limit.

Using a limiter alone with linear amplification is inadequate in hearing aids, since hearing loss is often level-dependent because of recruitment. Thus, hearing aids typically employ *automatic gain control* (AGC) which enables the *level compression* of sound – the amplification of low input levels more than high ones. Consequently, the dynamic range of the input signal is reduced so that it better fits the decreased dynamic range of the impaired hearing.

Figure 19.9 shows an example of an amplification curve that contains a noise gate, a range of linear amplification, compression, and limiting of the highest levels. The noise gate sets the lowest level of sound that is reproduced, thus preventing the amplification of system noise. Compression enables the overall gain to be increased so that low-level sound becomes audible without high-level sound becoming too loud. In addition, applying compression of high levels together with a limiter enables limiting the output level with less distortion than with just a simple limiter. Other strategies for range compression also exist, and these are reviewed by Dillon (2012).

Automatic gain control can be implemented with a feedback loop, as shown in Figure 19.10a. Figure 19.11 shows schematically how the time-domain output signal changes when the input signal is linearly amplified, limited, and compressed. The analysis conducted on the signal always has a finite integration time that the circuit takes to change the gain, as is shown in Figure 19.11.

Automatic gain control can also be implemented with a feedforward loop, as shown in Figure 19.10b. The gain applied to the signal is thus set by analysing either the input or output signals of the amplifier. The difference between the feedback and feedforward loops depends on other processing in the system. For example, if the device has a volume control, the positioning of the loop with respect to the volume control affects the result. A feedforward loop

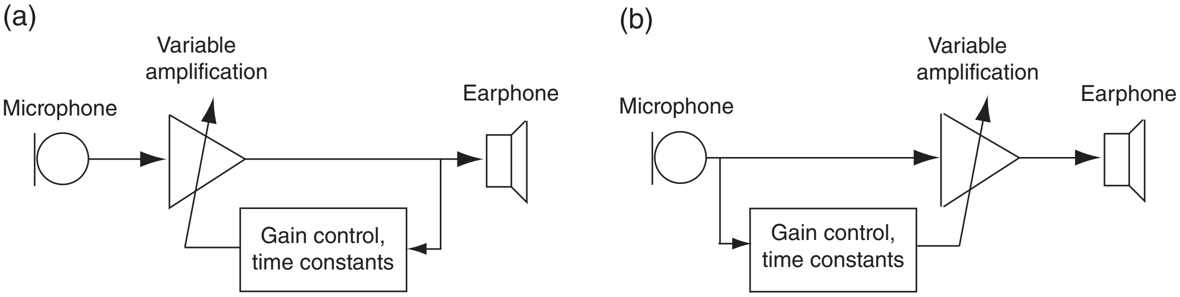

Figure 19.10 Automatic gain control (AGC) implemented with (a) a feedback loop, and (b) a feedforward loop.

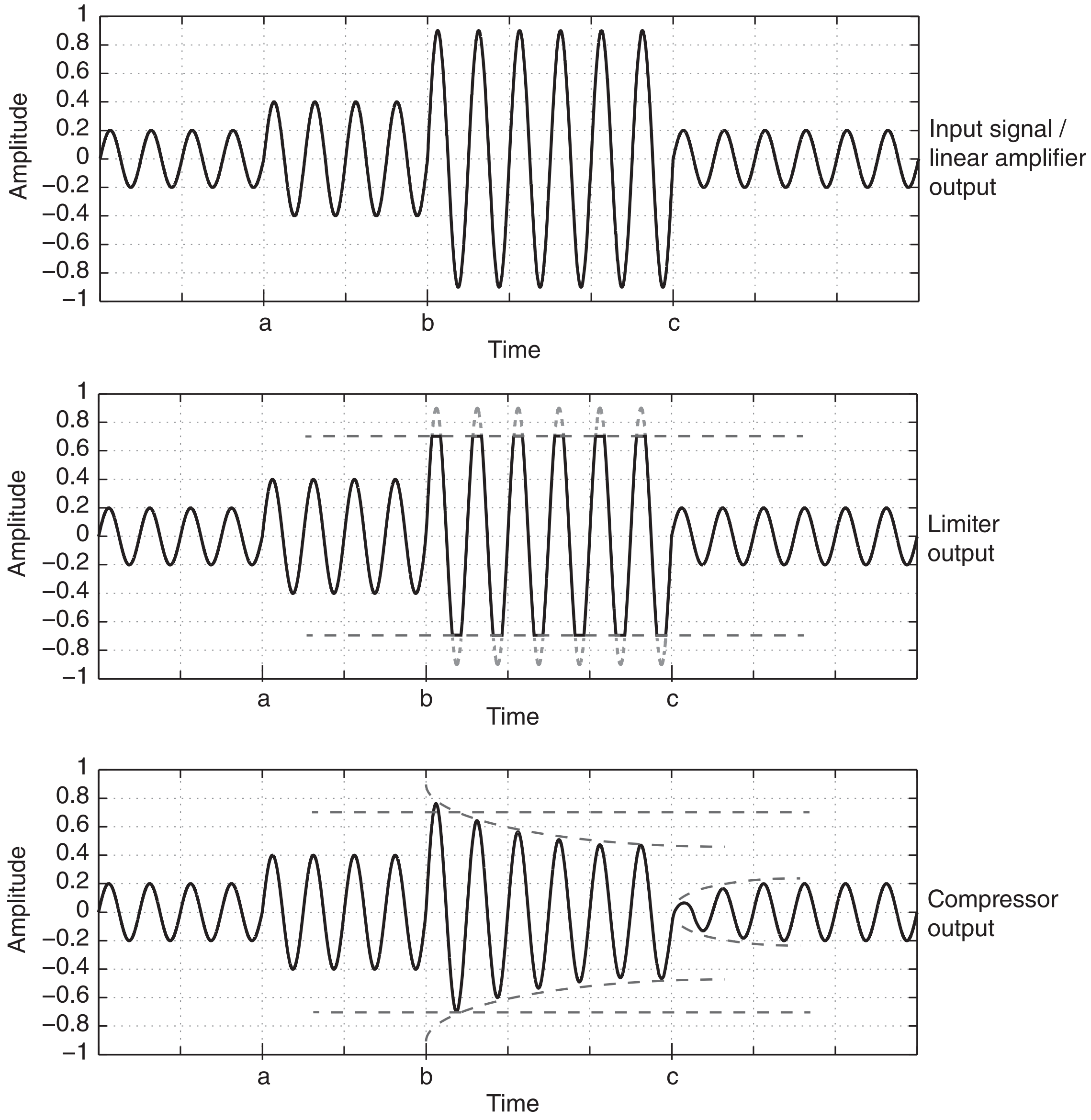

Figure 19.11 The effect of the amplifier implementation on the output level. In the interval between times b and c, the output signal is limited (middle panel) or compressed (the lowest panel). Courtesy of Teemu Koski.

enables global control of the volume without affecting the behaviour of compression, and a feedback loop after the volume control affects the volume only at low input levels and causes a greater proportion of sounds to be compressed heavily and limited.

Directional microphone systems

One approach to increasing the SNR in hearing aids is to use a *directional microphone* to amplify sound in some directions more than others. A simple and widely used design is a first-order subtractive directional microphone, which can be constructed with a single microphone having two ports or with two separate microphones, one port at each. Such a design makes possible first-order directivity patterns including cardioid, hyper-cardioid, super-cardioid, and dipole patterns, as in Figure 14.5 on page 290. In the one-microphone design, sound is fed to both sides of the microphone diaphragm, and the directivity pattern is controlled by the port spacing and an internal acoustic delay. In the two-microphone construction, the signals of the two microphones are subtracted, and the directivity pattern is controlled by the microphone spacing and an electrical delay between the microphone signals. By combining signals from a larger number of microphones, even more directive beams can be achieved. In general, the generation of a directive microphone beam by combining signals from several microphones is called *beamforming*.

It is sensible to generate the directivity pattern so that the direction of highest sensitivity is in the front, since in practical situations people generally look in the direction they want to listen to. Some beamforming algorithms are adaptive, so that the directivity pattern changes depending on the situation in order to minimize the level of unwanted noise. They are typically designed to adaptively set the lowest sensitivity in the direction of the most dominant noise source.

Other signal processing features

In addition to directional microphone systems, the SNR in the hearing aid output can be increased by single-channel noise reduction schemes (Dillon, 2012). These techniques use the temporal, spectral, and statistical information of the incoming sound to suppress the noise. Various *speech enhancement* algorithms assume some properties for speech and use these properties to attenuate unwanted noise and preserve speech-like signals.

Feedback cancellation helps to suppress the loud whistling caused by possible acoustic feedback from the hearing aid earphone to the hearing aid microphone. Feedback cancellation can be implemented, for example, with an adaptive filter that detects the feedback and adjusts the filter response adaptively to attenuate the feedback.

Binaural processing in hearing aids can be advantageous compared to monaural processing. In binaural processing, there is interaction between the left and right hearing aids in order to preserve or enhance the spatial hearing abilities of the user. For example, at high frequencies compression may suppress the sound on the ipsilateral side and leave it untouched on the contralateral side. This biases the ILD cue towards the median plane, which is undesirable. Binaural processing enables, for example, the compression gain to be made equal on both sides and interaural cues to be preserved, which further enhances the spatial hearing performance with hearing aids. Improvements provided with binaural processing have been reported, for example, in terms of speech intelligibility in conditions of noise; see, for example, Moore (2007) for a review. The input from multiple microphones can also be processed non-linearly, combined together with single-microphone processing, to achieve noise reduction or beamforming in hearing aids (Ahonen *et al.*, 2012; Hersbach *et al.*, 2013; Van den Bogaert *et al.*, 2009).

19.5.3 Transmission Systems and Assistive Listening Devices

Hearing aids often have some means to receive sound from an external source. The general aim of such devices is usually to suppress unwanted sounds and make target signals clear and audible, thus enhancing intelligibility even in difficult listening scenarios.

The hearing aid device may also receive sound wirelessly if it is equipped with an induction coil or a wireless receiver. Correspondingly, some public sites are equipped with an induction loop system which can be used to transmit sound wirelessly to the induction coil. Hearing aids may also receive sound via radio-frequency transmission. Various assistive listening devices utilize this channel. For example, an external microphone with a wireless transmitter can be useful when communicating in noisy environments if the microphone can be placed close to the target sound source to maximize the SNR in the hearing aid of the user. Assistive listening devices can also provide the means to connect to other communication devices, such as telephones.

19.6 Implantable Hearing Solutions

19.6.1 Cochlear Implants

A *cochlear implant* (Clark, 2003; Zeng *et al.*, 2003) is a surgically implanted electrical device that transmits sound to the auditory nerve fibres in the cochlea. The conductive part of the auditory path and part of the cochlea are bypassed. A cochlear implant can provide a usable sense of hearing in those cases where the hearing impairment is so severe that a hearing aid cannot provide adequate help. These cases include deafness or severe to profound bilateral hearing loss due to sensorineural impairment.

Figure 19.12 shows a system diagram of a cochlear implant. Microphones and the sound processor are located typically behind the ear, in a similar manner to BTE hearing aids. The sound processor typically employs similar signal processing schemes as in hearing aids, such as beamforming and noise management. After this pre-processing, the sound processor divides the incoming sound into several frequency channels and generates a pulse-train signal representing each channel. Several pulse-coding methods have been developed to implement this (Loizou, 1999). The pulse-train signal is routed to the transmitter coil, which is located on the scalp, behind the ear. The receiver coil of the implant, which receives the signal via induction, is at the same location but under the skin. From the receiver coil, the signal is fed to the electrode array. The electrode array is placed inside the cochlea with a surgical operation that requires high precision. The electrode contacts in the array stimulate the auditory nerve fibres, aiming to trigger patterns of activity in the auditory nerve that imitate the inputs from the hair cells of a functional basilar membrane. Different channels are fed to different electrode contacts, which are placed at different locations on the basilar membrane, thus providing a rough place-code for different frequencies.

The first cochlear implants utilized only one frequency channel. These devices were not able to provide frequency discrimination, but only some sense of the general temporal structure of the signal. Modern cochlear implants employ electrode arrays with multiple frequency channels. However, the cross talk between the channels is a significant factor which limits the frequency resolution that can be achieved with cochlear implants (Schnupp *et al.*, 2011).

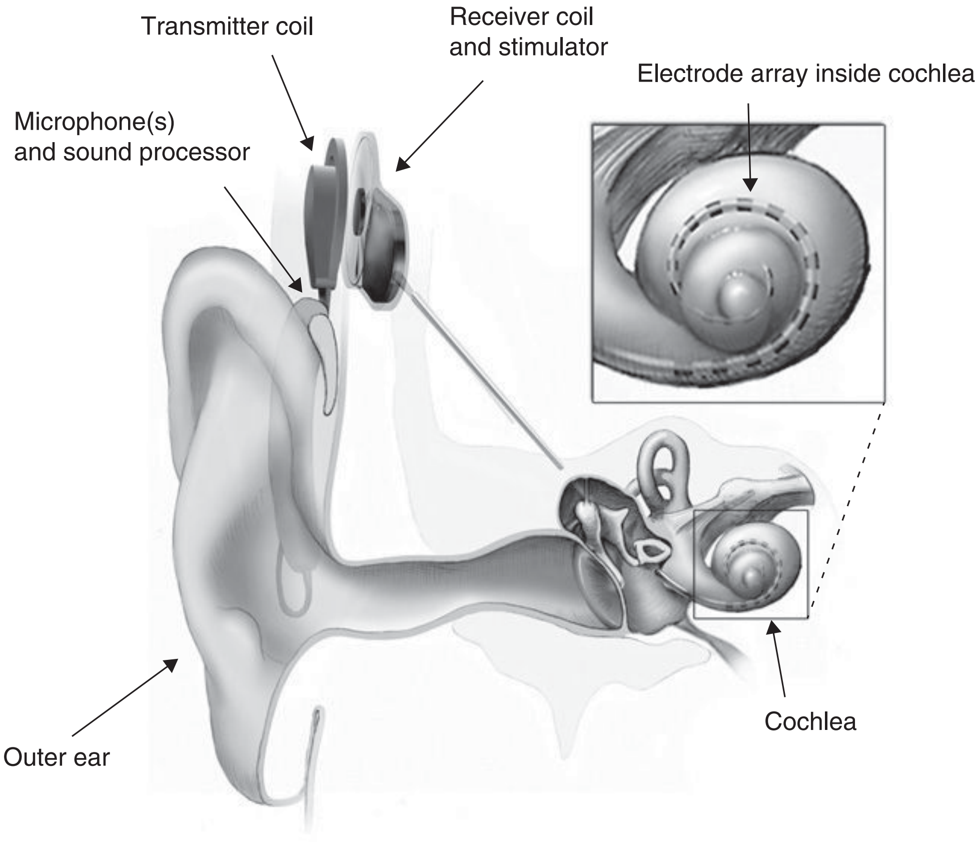

Figure 19.12 System diagram of a cochlear implant. Adapted from National Institutes of Health (2014).

The implication is that the impulses from a single electrode are received by a frequency region of the cochlea larger than one critical band.

By December 2012, approximately 324 200 cochlear implants had been fitted worldwide (National Institutes of Health, 2014). Hence, extensive knowledge has been gained about the outcomes of cochlear implantation. There is individual variation on the hearing performance achieved with cochlear implants. With modern implants and successful rehabilitation, most patients achieve the ability to have a telephone conversation (Zeng, 2004). Visual cues, such as lip-reading, can enhance the communication performance significantly. Even in cases where good speech communication abilities are not achieved, the increased auditory awareness provided by the implant can be advantageous.

The ability to perceive pitch is immensely poorer than that in normal hearing, which naturally makes musical structures such as melody, harmony, and the bass line very difficult if not impossible to perceive. Consequently, the desire to listen to music varies considerably among cochlear implant users. Some users like to listen to music even though it is distorted to the extent that it is difficult to discriminate the instruments from each other. Many instruments producing harmonic tones are generally perceived as unpleasantly distorted, but the basic rhythm of music can be heard almost normally.

In adults, cochlear implants are fitted mainly to those individuals who already spoke before their hearing loss. In adults who were born deaf, the results of cochlear implantation have not been optimal: with no previous sound-evoked input, the adult auditory system may be unable to interpret auditory input and may not have the plasticity to learn it. In contrast, a cochlear implant is well suited for deaf infants – if received at an early enough age – since even unilateral cochlear implantation enables the auditory system to develop to such an extent that spoken language skills are often at age-appropriate levels (Nicholas and Geers, 2007).

19.6.2 Electric-Acoustic Stimulation

Electric-acoustic stimulation refers to electric and acoustic hearing – a cochlear implant with or without an ipsilateral hearing aid. Modern atraumatic implant electrode arrays and so-called soft surgical techniques have resulted in preserved residual hearing in the implanted ear, typically in the low-frequency region (Adunka *et al.*, 2013; Skarzynski *et al.*, 2014). Consequently, in electric-acoustic stimulation, the cochlear implant provides high frequencies and the hearing aid the low frequencies. This combination can be beneficial to patients who have high-frequency hearing loss to such a degree that it cannot be successfully compensated for with a hearing aid, but have an adequate degree of residual low-frequency hearing that can benefit from acoustic hearing. Electric-acoustic stimulation can provide not only better speech intelligibility than a cochlear implant alone, but also enhanced pitch perception and music appreciation (Gantz *et al.*, 2005).

19.6.3 Bone-Anchored Hearing Aids

A *bone-anchored hearing aid* (BAHA) differs in its operating principle from the hearing aid types discussed earlier in this book. In a BAHA system, a small implant is inserted into the skull behind the ear. The sound processor is attached to this implant, and the implant feeds the sound to the skull as mechanical vibrations. Consequently, sound travels to the inner ear via bone conduction. BAHA are suitable for individuals with an impaired outer or middle ear, since these parts are bypassed by the bone-conduction path. BAHA systems can provide better speech intelligibility compared to conventional hearing aids, especially if the air-conduction hearing thresholds differ considerably from the bone-conduction hearing thresholds (de Wolf *et al.*, 2011).

19.6.4 Middle-Ear Implants

In *middle-ear implants*, a transducer is attached to the auditory ossicles or to the round window that vibrates the structures in the same manner that air-conducted sound would. Middle-ear implant systems vary in their way of picking up incoming sound. One approach is to use an external sound processor and an inductive link, similarly to cochlear implants. The microphone can also be implanted under the skin or placed in the ear canal. Reviews by Butler *et al.* (2013) and Kahue *et al.* (2014) conclude that middle-ear implants, on average, provide similar benefits in terms of speech intelligibility to conventional hearing aids. On the other hand, middle-ear implants have several other benefits compared to hearing aids. According to the review by Dillon (2012), the benefits are, to mention just a few, an extended high-frequency response, lower distortion, increased gain before feedback occurs, and benefits due to no obstruction of the ear canal. Furthermore, coupling to the round window can provide usable hearing to patients with a dysfunctional middle-ear system (Colletti *et al.*, 2006).

Summary

This chapter presented a general overview of technical audiology. The chapter discussed the causes and symptoms of hearing impairments, how they are assessed with audiometry, and what solutions have been developed to manage them with hearing aids and implantable hearing devices. Since the topic is broad, the interested reader is encouraged to explore the relevant literature for more in-depth information.

Further Reading

A good amount of literature exists in the field of technical audiology. Sensorineural hearing loss and its perceptual effects are discussed by Moore (2007). A general introduction to audiology is made by Martin and Clark (2006) and to hearing aids by Dillon (2012). An overview of signal processing methods for hearing aids can be found in Kates (1998). For a review of cochlear implant research, see, for example, Wilson and Dorman (2008) and Zeng *et al.* (2003).

References

Adunka, O.F., Dillon, M.T., Adunka, M.C., King, E.R., Pillsbury, H.C., and Buchman, C.A. (2013) Hearing preservation and speech perception outcomes with electric-acoustic stimulation after 12 months of listening experience. *Laryngoscope*, **123**(10), 2509–2515.

Ahonen, J., Sivonen, V., and Pulkki, V. (2012) Parametric spatial sound processing applied to bilateral hearing aids. *45th Int. Conf. Audio Eng. Soc.* AES.

Butler, C.L., Thavaneswaran, P., and Lee, I.H. (2013) Efficacy of the active middle-ear implant in patients with sensorineural hearing loss. *J. Laryng. & Otol.* **127**(Suppl 2), 8–16.

Campbell, K.C.M. and Mullin, G. (2012) Impedance audiometry. *Medscape reference*. `http://emedicine.medscape.com/article/1831254-overview`.

Clark, G. (2003) *Cochlear Implants: Fundamentals and applications*. Springer.

Colletti, V., Soli, S.D., Carner, M., and Colletti, L. (2006) Treatment of mixed hearing losses via implantation of a vibratory transducer on the round window. *Int. J. Audiol.*, **45**(10), 600–608.

de Wolf, M.J., Hendrix, S., Cremers, C.W., and Snik, A.F. (2011) Better performance with bone-anchored hearing aid than acoustic devices in patients with severe air–bone gap. *Laryngoscope*. **121**(3), 613–616.

Dillon, H. (2012) *Hearing Aids*. Boomerang Press.

European working group on genetics of hearing impairment (1996) Info letter II. `http://audiology.unife.it/ www.gendeaf.org/hear/infoletters/Info_02.PDF`.

Favrot, S. and Buchholz, J.M. (2010) LoRA: A loudspeaker-based room auralization system. *Acta Acustica united with Acustica*, **96**, 364–375.

Gantz, B.J., Turner, C., Gfeller, K.E., and Lowder, M.W. (2005) Preservation of hearing in cochlear implant surgery: advantages of combined electrical and acoustical speech processing. *Laryngoscope*, **115**(5), 796–802.

Hersbach, A.A., Grayden, D.B., Fallon, J.B., and McDermott, H.J. (2013) A beamformer post-filter for cochlear implant noise reduction. *J. Acoust. Soc. Am.*, **133**(4), 2412–2420.

ISO 389–1 (1998) Acoustics standard. Reference zero for the calibration of audiometric equipment.

ISO 8253-1 (2010) Acoustics standard. Audiometric testing methods. Part 1: Pure-tone air and bone conduction audiometry.

ISO 8253-2 (2009) Acoustics standard. Audiometric testing methods. Part 2: Sound field audiometry with pure-tone and narrow-band test signals.

ISO 8253-3 (1996) Acoustics standard. Audiometric testing methods. Part 3: Speech audiometry.

Jastreboff, P.J. (1990) Phantom auditory perception (tinnitus): mechanisms of generation and perception. *Neurosci. Res.* **8**(4), 221–254.

Jastreboff, P.J., Gray, W.C., and Gold, S.L. (1996) Neurophysiological approach to tinnitus patients. *Am. J. Oto.* **17**(2), 236–240.

Kahue, C.N., Carlson, M.L., Daugherty, J.A., Haynes, D.S., and Glasscock, M.E. III (2014) Middle ear implants for rehabilitation of sensorineural hearing loss: A systematic review of FDA approved devices. *Otol. Neurotol.*.

Kates, J.M. (1998) *Signal Processing for Hearing Aids*. Kluwer Academic Publishers.

Kollmeier, B. and Wesselkamp, M. (1997) Development and evaluation of a German sentence test for objective and subjective speech intelligibility assessment. *J. Acoust. Soc. Am.*, **102**(4), 2412–2421.

Koski, T., Sivonen, V., and Pulkki, V. (2013) Measuring speech intelligibility in noisy environments reproduced with parametric spatial audio. *Proc. 135th Conv. Audio Eng. Soc.*.

Liberman, M.C. and Dodds, L.W. (1984) Single-neuron labeling and chronic cochlear pathology. III. Stereocilia damage and alterations of threshold tuning curves. *Hearing Res.*, **16**(1), 55–74.

Loizou, P.C. (1999) Signal-processing techniques for cochlear implants. *IEEE Trans. Engineering in Medicine and Biology* (May/June), 34–46.

Martin, F.N. and Clark, J.G. (2006) *Introduction to Audiology*. Prentice-Hall.

Miller, J.D. (1974) Effects of noise on people. *J. Acoust. Soc. Am.*, **56**(3), 729–764.

Minnaar, P., Frølund, A.S., Simonsen, C.S., Søndersted, B., Oakley, S., Dalgas, A., and Bennedbæk, J. (2013) Reproducing real-life listening situations in the laboratory for testing hearing aids. *Proc. 135th Conv. Audio Eng. Soc.* AES.

Moore, B.C. (2007) *Cochlear Hearing Loss: Physiological, psychological and technical issues*. John Wiley & Sons.

National Institutes of Health (2014) Nih publication no. 11-4798. `http://www.nidcd.nih.gov/health/hearing/pages/coch.aspx`.

Nicholas, J.G. and Geers, A.E. (2007) Will they catch up? The role of age at cochlear implantation in the spoken language development of children with severe to profound hearing loss. *J. Speech, Lang. Hearing Res.*, **50**(4), 1048–1062.

Nilsson, M., Soli, S.D., and Sullivan, J.A. (1994) Development of the hearing in noise test for the measurement of speech reception thresholds in quiet and in noise. *J. Acoust. Soc. Am.*, **95**(2), 1085–1099.

Schnupp, J., Nelken, I., and King, A. (2011) *Auditory Neuroscience: Making sense of sound*. MIT Press.

Seeber, B.U., Kerber, S., and Hafter, E.R. (2010) A system to simulate and reproduce audio-visual environments for spatial hearing research. *Hearing Res.* **260**, 1–10.

Skarzynski, H., Lorens, A., Matusiak, M., Porowski, M., Skarzynski, P.H., and James, C.J. (2014) Cochlear implantation with the nucleus slim straight electrode in subjects with residual low-frequency hearing. *Ear and Hearing*, **35**(2), 33–43.

Toivanen, J. (1976) *Teknillinen akustiikka*. Otakustantamo, Espoo.

Toppila, E. (2000) *A systems approach to individual hearing conservation*. PhD thesis, University of Helsinki.

Van den Bogaert, T., Doclo, S., Wouters, J., and Moonen, M. (2009) Speech enhancement with multichannel Wiener filter techniques in multimicrophone binaural hearing aids. *J. Acoust. Soc. Am.*, **125**(1), 360–371.

WHO (2013) Fact sheet n 300: Deafness and hearing loss. `http://www.who.int/mediacentre/factsheets/fs300/en/`.

Wilson, B.S. and Dorman, M.F. (2008) Cochlear implants: A remarkable past and a brilliant future. *Hearing Res.*, **242**(1–2), 3–21.

Zeng, F.G. (2004) Trends in cochlear implants. *Trends in Amplif.*, **8**(1), 1–34.

Zeng, F.G., Popper, A.N., and Fay, R.R. (2003) *Cochlear Implants: Auditory prostheses and electrical hearing*. Springer.