Springer Handbook of Auditory Research

Gerald R. Popelka
Brian C. J. Moore
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Arthur N. Popper *Editors*

Hearing Aids

ASA Press

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Series Preface

The following preface is the one that we published in Volume 1 of the *Springer Handbook of Auditory Research* back in 1992. As anyone reading the original preface, or the many users of the series, will note, we have far exceeded our original expectation of eight volumes. Indeed, with books published to date, and those in the pipeline, we are now set for more than 50 volumes in *SHAR*, and we are still open to new and exciting ideas for additional books.

We are very proud that there seems to be consensus, at least among our friends and colleagues, that *SHAR* has become an important and influential part of the auditory literature. While we have worked hard to develop and maintain the quality and value of *SHAR*, the real value of the books is very much because of the numerous authors who have given their time to write outstanding chapters and to our many coeditors who have provided the intellectual leadership to the individual volumes. We have worked with a remarkable and wonderful group of people, many of whom have become great personal friends of both of us. We also continue to work with a spectacular group of editors at Springer. Indeed, several of our past editors have moved on in the publishing world to become senior executives. To our delight, this includes the current president of Springer US, Dr. William Curtis.

But the truth is that the series would and could not be possible without the support of our families, and we want to take this opportunity to dedicate all of the *SHAR* books, past and future, to them. Our wives, Catherine Fay and Helen Popper, and our children, Michelle Popper Levit, Melissa Popper Levinsohn, Christian Fay, and Amanda Fay, have been immensely patient as we developed and worked on this series. We thank them, and state, without doubt, that this series could not have happened without them. We also dedicate the future of *SHAR* to our next generation of (potential) auditory researchers—our grandchildren—Ethan and Sophie Levinsohn; Emma Levit; and Nathaniel, Evan, and Stella Fay.

Preface 1992

The *Springer Handbook of Auditory Research* presents a series of comprehensive and synthetic reviews of the fundamental topics in modern auditory research. The volumes are aimed at all individuals with interests in hearing research including advanced graduate students, postdoctoral researchers, and clinical investigators. The volumes are intended to introduce new investigators to important aspects of hearing science and to help established investigators to better understand the fundamental theories and data in fields of hearing that they may not normally follow closely.

Each volume presents a particular topic comprehensively, and each serves as a synthetic overview and guide to the literature. As such, the chapters present neither exhaustive data reviews nor original research that has not yet appeared in peer-reviewed journals. The volumes focus on topics that have developed a solid data and conceptual foundation rather than on those for which a literature is only beginning to develop. New research areas will be covered on a timely basis in the series as they begin to mature.

Each volume in the series consists of a few substantial chapters on a particular topic. In some cases, the topics will be ones of traditional interest for which there is a substantial body of data and theory, such as auditory neuroanatomy (Vol. 1) and neurophysiology (Vol. 2). Other volumes in the series deal with topics that have begun to mature more recently, such as development, plasticity, and computational models of neural processing. In many cases, the series editors are joined by a coeditor having special expertise in the topic of the volume.

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Volume Preface

Hearing loss is a major health condition that affects a very large portion of the general population. All people with hearing loss as measured by the audiogram potentially would benefit from a conventional acoustic hearing aid that amplifies sounds to compensate for the decrease in hearing sensitivity, although many such people do not regularly use hearing aids. This volume provides an overview of current key issues in hearing aid research from the perspective of many different disciplines. The volume offers insight into the scientific knowledge, current technology, and future technology that can help improve hearing aids. The book should prove useful to people with a wide range of backgrounds, including engineers, basic scientists, ENT specialists, and audiologists, as few people have expertise over the whole range of the individual disciplines that are relevant.

Chapter 1 by Moore and Popelka provides an overview of this volume as well as a discussion of general principles associated with hearing aids. In Chap. 2, Curan and Curan consider the incidence and causes of hearing loss. Chapter 3, by Killion, Van Halteren, Stenfelt, and Warren, describes the transducers used in hearing aids: these are the microphones that are used to pick up sounds and the receivers that are used to generate the sound after processing by the hearing aid circuitry. In Chap. 4, Launer, Zakis, and Moore describe the signal processing that is used in digital hearing aids to restore audibility while maintaining comfortable loudness, reducing the effects of background sounds, and increasing sound quality and listening comfort.

Chapter 5, by Mecklenburger and Groth, describes ways in which signals can be transmitted wirelessly from external devices (remote microphones, conventional and mobile telephones, televisions, stereos, computers, tablets) to hearing aids and between bilaterally fitted hearing aids. In Chap. 6, Souza describes the effects of hearing loss and of hearing aids on the perception of speech. Chapter 7, by Akeroyd and Whitmer, describes the influence of hearing loss and hearing aids on spatial perception, synthesizing material from a large number of published studies. In Chap. 8, Zakis describes the effect of hearing aids on the perception of music and considers the characteristics required of microphones, amplifiers, signal processors, and receivers to provide a relatively undistorted, high-fidelity, noise-free representation of musical signals.

Chapter 9, by Munro and Mueller, is concerned mainly with methods for fitting hearing aids and methods for verifying that the fitting is correct in terms of the sound delivered to the tympanic membrane. In Chap. 10, Whitmer, Wright-Whyte, Holman, and Akeroyd focus on validation of the performance of hearing aids, especially via the use of questionnaires. They also review the very large number of questionnaires that have been developed for the purpose of validation of hearing aid performance and describe the great variety of domains that have been addressed. Finally, Chap. 11, by Popelka and Moore, describes possible future directions for hearing aids and hearing aid research.

Much of the material in earlier *SHAR* volumes that provide basic science of hearing leads to this volume, where the authors describe the most common intervention for hearing loss. Hearing aids and other assistive devices have been discussed in *Speech Processing in the Auditory System* (Vol. 18, edited by Greenberg, Ainsworth, Popper, and Fay in 2004), *Cochlear Implants: Auditory Prostheses and Electric Hearing* (Vol. 20, edited by Zeng, Popper, and Fay in 2004), and *Auditory Prostheses: New Horizons* (Vol. 39, edited by Zeng, Popper, and Fay in 2011), as well as *The Middle Ear: Science, Otosurgery, and Technology* (Vol. 46, edited by Puria, Fay, and Popper in 2013).

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Chapter 1
Introduction to Hearing Aids

Brian C.J. Moore and Gerald R. Popelka

Abstract This chapter describes the background to the volume and introduces the range of disciplines that are involved in the development and evaluation of hearing aids. It then describes some basic aspects of hearing aids, such as the different styles of hearing aids and requirements for batteries. The chapter then gives an overview and brief summary of the remaining chapters in the volume, describing the components that are used in hearing aids; the needs of users; the signal processing that is used in hearing aids for listening to speech, music, and environmental sounds; wireless accessories and wireless communication between hearing aids; the fitting of hearing aids; the benefits of bilateral fittings; the verification of fittings; and evaluation of effectiveness.

Keywords Batteries • Epidemiology • Fitting methods • Hearing aids • Hearing impairment • Microphones • Music perception • Real-ear measurements • Receivers • Signal processing • Spatial perception • Speech perception • Validation questionnaires • Wireless connectivity

1.1 Introduction

Hearing loss is a major health condition that affects a very large portion of the general population and has a very wide range of etiologies. Most auditory pathologies result in a decrease in hearing sensitivity, although there has been increasing interest recently in "hidden hearing loss," in which hearing difficulties may be experienced without any abnormality in the audiogram (Kujawa and Liberman 2009; Füllgrabe et al. 2015). Actual hearing restoration is possible with certain surgical or

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pharmaceutical interventions in a small minority of cases. Partial functional hearing restoration is possible with other surgical interventions, and the promise of hearing restoration resulting from biological and pharmaceutical interventions is on the distant horizon (Oshima et al. 2010; Rivolta 2013). However, these hearing restoration interventions apply to a very small portion of the entire hearing-impaired population and/or have yet to be developed. All people with a nontreatable hearing loss as measured by the audiogram potentially would benefit from a conventional acoustic hearing aid that amplifies sounds to compensate for the decrease in hearing sensitivity, although in practice many such people do not use hearing aids; see Chap. 2 by Curhan and Curhan.

As described in Chap. 2, the largest proportion of hearing impairment in the population is due to permanent sensorineural hearing loss associated with the aging process. The prevalence of age-related permanent sensorineural hearing loss is increasing because of the increase in longevity and the rise in the population in many countries associated with the increased birth rates that occurred after World War II. For the next generation, well over 1 billion people in the world will have age-related, permanent sensorineural hearing loss for which acoustic hearing aids are the only option.

Hearing aids are extremely complicated electroacoustic devices that must operate in hostile environments, both physically (heat, cold, humidity, water, electric and magnetic fields, wind) and acoustically (very large range of input sound levels and frequencies), with many inherent restrictions, including limited power supply, the need to be very small, and the proximity of the microphone to the receiver (the name for the loudspeaker that actually generates the amplified sound). It should be no surprise that these devices require a large cadre of professionals from a wide variety of wholly independent disciplines to conceive of, design, manufacturer, test, distribute, fit, adjust, and evaluate hearing aids. These professionals include physicians, audiologists, engineers, psychologists, psychoacousticians, acousticians, experts in ergonomics, and many others from an equally wide range of institutions varying from clinical facilities; small, medium, and large companies; charities; various government institutions; and universities. This multidisciplinary nature of hearing aids is one of the most challenging aspects of hearing aid research. Very few scientific educational degree programs provide a systematic introduction to all of the complex issues relating to hearing aids, resulting in challenges and obstacles that limit or prevent effective research. Few if any individuals are knowledgeable about all of the factors relevant to hearing aids. However, effective research is the key to the conduct of high-impact studies that ultimately result in significant clinical advances and improved patient outcomes.

This volume provides an overview of current key issues in hearing aid research from the perspective of many different disciplines, not only those of the key funding agencies but also those of the scientists and clinicians who are currently involved in hearing aid research. The volume offers insights into experience with hearing aids, factors affecting the candidacy for and efficacy of hearing aids, perceptual factors, current technology and future technology, and the interaction of these variables.

These insights can make scientists and clinicians aware of the important issues over the entire range of the individual disciplines that are relevant, paving the way for future research to improve hearing aids.

This book should be regarded as complementing existing books on hearing aids. The book *Digital Hearing Aids* by Kates (2008) has a particular focus on the signal processing that is performed in hearing aids, and, for the technically minded, provides a good understanding of the operation of multichannel compression, noise reduction, and acoustic feedback reduction. The book *Hearing Aids* by Dillon (2012) and the series *Modern Hearing Aids* (Bentler and Mueller 2013; Mueller et al. 2013) are aimed more at clinicians and include many practical tips and recommendations.

1.2 Population

Chapter 2 reviews the incidence of hearing loss and the many factors that can contribute to the development of hearing loss. The authors point out that hearing loss affects about 20 % of individuals in at least one ear—a higher prevalence than any other sensory disorder. Remarkably, only about 20 % of people who could potentially benefit from hearing aids actually use them. The etiology of hearing impairment reflects the cumulative and interacting influences of many factors, including aging, genetics, epigenetics, environment (e.g., exposure to noise at work or during leisure activities, or exposure to industrial solvents), health, diet, and lifestyle. This makes it difficult to determine the specific factors contributing to hearing loss in any given individual as well as the anatomical and physiological changes associated with the hearing loss. Work is ongoing to identify risk factors for hearing impairment. This may lead to a better understanding of the cellular and molecular mechanisms associated with acquired hearing impairment and could contribute to efforts toward prevention; early detection; delay of progression; and medical, pharmaceutical, surgical, and biological interventions to restore hearing.

1.3 Technical Aspects of Hearing Aids

1.3.1 Components of Hearing Aids

Most modern hearing aids contain the following components (see Chap. 4 by Launer, Zakis, and Moore and Chap. 8 by Zakis for block diagrams):

1. One or more microphones to pick up the external sound (described in detail in Chap. 3 by Killion, Van Halteren, Stenfelt, and Warren).

2. A preamplifier for each microphone to increase the electrical signal magnitude combined with a low-pass filter to limit the highest frequency that is present. Usually, this has a cutoff frequency of 9 kHz or lower.
3. An analog-to-digital converter for each amplified microphone signal. This converts the continuously varying voltage to a series of numbers representing the voltage at discrete regularly spaced times. The number of samples per second is called the sampling rate. This must be two or more times greater than the cutoff frequency of the low-pass filter described in (2) (Kates 2008). Typically, there might be 20,000 samples per second, allowing the representation of frequencies up to about 9,000 Hz. The voltage of each sample is usually represented with 16-bit precision, which means that the largest voltage that can be represented is 65,536 (2^{16}) times the smallest voltage that can be represented. This corresponds to a dynamic range of about 96 dB.
4. A digital signal processor, essentially a miniature computer, that performs operations such as frequency-dependent amplification, amplitude compression and limiting, noise reduction, cancellation of acoustic feedback (the squealing resulting from the receiver output getting back to the microphones of the aid), and directional processing (described in detail in Chap. 4).
5. A receiver, which is a device for converting the output of the signal processor to sound. Sometimes this is accomplished via a digital-to-analog converter, but in many hearing aids the digital output of the processor is converted directly into sound by the receiver (see Chap. 3).
6. A battery for powering the circuitry and the receiver.
7. A casing in which most of the aforementioned components are housed, often customized to fit into the convoluted external ear and ear canal.

1.3.2 Styles of Hearing Aids

Hearing aids are made in several different styles, as illustrated in Fig. 1.1. One style employs a single case that contains all of the components and is referred to as either in-the-ear (ITE) if the case is visible in the concha (the bowl of the pinna; Fig. 1.1a) or completely-in-the-canal (CIC) if the case is small enough to fit completely in the ear canal (Fig. 1.1b). Some CIC aids can fit more deeply into the ear canal than shown in Fig. 1.1b and are essentially invisible. The most prevalent style currently is the behind-the-ear (BTE) aid (Fig. 1.1c) in which most of the components are housed in a small case that fits behind the pinna and the microphones are positioned just above the pinna (Kochkin 2010). The receiver can be mounted in the BTE case with the sound delivered to the ear canal by an acoustic tube fitted into a custom-made earmold or held in place via a soft "dome." Alternatively, the receiver may be placed in the ear canal, held in place via a dome, and connected via a very thin wire to the BTE part. This BTE style is referred to as receiver in the canal (RIC) or receiver in the ear (RITE) and is illustrated in Fig. 1.1d.

Fig. 1.1 Illustration of some styles of hearing aids. (**a**) In the ear. (**b**) Completely in the canal. (**c**) Behind the ear. (**d**) Behind the ear with receiver in the canal and open dome

1.3.3 Closed and Open Fittings

The earmold or dome can be sealed into the ear canal; this is called a closed fitting. A closed fitting may be required when the hearing loss is severe or profound and considerable gain is needed. The seal prevents intense amplified sound in the ear canal from being perceived by others and it also helps to control or limit acoustic feedback. However, for people with mild or moderate hearing loss at low frequencies, a closed fitting can give rise to the "occlusion effect," whereby the user's own voice sounds loud and boomy (Killion et al. 1988). This happens because the sound of the user's voice is transmitted into the ear canal via the bones of the head and cannot escape because of the blocked ear canal (Stone et al. 2014). An earmold or dome with a large vent or opening (Fig. 1.1d), called an open fitting, can reduce or avoid the occlusion effect and allow much of the external sound to enter the ear canal unamplified for frequency regions where amplification is not required. Open fittings have become very popular in recent years because they are physically comfortable, may be less visible than the large ITE styles, and do not require a custom earmold to be made, so a person can be fitted with a hearing aid on his or her first visit to the clinic. In some cases, an open fitting may prevent exacerbation of certain medical conditions in the external ear canal.

1.3.4 Batteries

Hearing aid batteries need to be small but to have sufficient capacity to power both the circuitry (preamplifier, analog-to-digital converter, signal processor, and for some aids the wireless receiver and transmitter) and the receiver. Most hearing aid batteries are disposable, have a single-cell structure, and generate approximately 1.5 V when "fresh." Hearing aids are designed to work properly when the voltage drops to a value as low as 1.1 or 1.0 V. Many hearing aids generate an audible warning via the receiver when the battery is running low but before it is completely depleted to alert the user of the need to change the battery. Common battery sizes for hearing aids are 675 (the largest, mainly used in large BTE devices), 312, 13, and 10 (the smallest, used mainly in very small BTE devices or ITE devices). A small-to-medium battery, such as the 312, typically has a capacity of 180 mAh, so that it can generate, for example, a current of 1 mA for 180 h, corresponding to an operational life of 6–10 days.

The most common type of hearing aid battery is the disposable zinc–air battery, which generates power by oxidizing zinc using oxygen from the air. Zinc–air batteries have high energy densities and are relatively inexpensive to produce. The high energy density is possible because the oxygen used in the reaction comes from the air and does not need to be part of the battery cell. New zinc–air batteries are supplied with a small tabbed seal that covers an opening in the cell casing. This tabbed seal prevents activation of the oxidation process so the battery can be stored for up to 3 years. When the tab is removed, air enters the casing and the battery becomes fully activated within several seconds or tens of seconds. The battery then starts to run down whether or not it is actually used in a hearing aid. The output voltage remains quite stable until the cell approaches exhaustion, which occurs between 1 and 10 days of use. The high range of durations of use occurs because of the large range of battery capacities and the large range of power requirements of hearing aids. Hearing aids that transmit and receive electromagnetic signals, as described in Chaps. 4 and 5 (by Mecklenburger and Groth), tend to consume more power than those that do not. The short-term demand for current is also higher for hearing aids that transmit and receive electromagnetic signals than for those that do not, which means that, for the former, the battery must have a low internal resistance. Some other factors that affect battery life are discussed in Chap. 8. In a survey, 18 % of hearing aid users were dissatisfied with the operational life of their hearing aid batteries (Kochkin 2010).

Some hearing aid users, especially older individuals and those with limited manual dexterity, find it difficult to change the battery on a hearing aid; it is a fiddly job, and it is easy to drop the battery and lose it. One solution to this problem is the use of rechargeable batteries. Typically, when a hearing aid is supplied with a rechargeable battery, the battery cannot be removed from the hearing aid and is recharged by placing the aid in a small inductive charging station. This is usually found to be easy, even by those with limited manual dexterity. The batteries typically last for about 20 h and require 2–4 h to be recharged. Most users recharge the battery overnight.

It is possible to buy rechargeable batteries for use as an alternative to zinc–air batteries. These have to be removed from the hearing aid for charging so they do not solve problems associated with limited manual dexterity. The main advantage is a saving in cost. Although a rechargeable battery costs more than a disposable zinc–air battery and a separate charger has to be purchased, the rechargeable battery can be used and recharged many times so there is eventually a cost saving. Most rechargeable batteries intended for conventional hearing aids use nickel–metal–hydride technology and are housed in a stainless steel casing the size of a conventional disposable battery. Rechargeable batteries are generally more environmentally friendly than disposable batteries.

1.3.5 Transducers

Chapter 3 describes the transducers used in hearing aids: these are the microphones that are used to pick up sounds and the receivers that are used to generate the sound after processing by the hearing aid circuitry. Both microphones and receivers have changed remarkably since the 1950s, becoming smaller and more efficient. The microphones must respond over a wide frequency range and have low internal noise despite their small size. This is challenging, but, in principle, the problem has been solved. As stated by Killion, Van Halteren, Stenfelt, and Warren (Chap. 3), "Modern hearing aid transducers have virtually eliminated the bandwidth and response limitations of the past: microphones and receivers with 16-kHz bandwidth and high-fidelity response are now available." Despite this, most hearing aids currently on the market do not provide useful amplification for frequencies above about 5 kHz (Moore et al. 2001; Aazh et al. 2012). Furthermore, most manufacturers of hearing aids incorporate low-level expansion (gain reduction when the signal level decreases below a certain value) to prevent microphone noise and circuit noise from being audible. This can lead to reduced intelligibility for weak speech sounds (Moore et al. 2004; Plyler et al. 2007).

An important issue in hearing aid design is the dynamic range of the input. This refers to the very large range of sound levels that can be encountered in everyday life. To restore hearing to "normal" for a person with a hearing loss would require that sounds with levels close to 0 dB sound pressure level (SPL) were above the low-level noise inherent in the microphone and analog-to-digital converter. At the other extreme, peak levels of 116 dB SPL or more can occur from a variety of sources, such as at live concerts. Although in principle hearing aid microphones can handle this wide dynamic range of 116 dB, in practice the microphone preamplifier will often clip or saturate for input sound levels above about 110 dB SPL because of limitations in the voltage supplied by the hearing aid battery. In addition, as mentioned earlier, the analog-to-digital converters used in hearing aids typically have 16-bit resolution, which gives a maximum dynamic range of only approximately 96 dB. As a result, some hearing aids produce annoying distortion when listening to higher level sounds such as live music (Madsen and Moore 2014). Some methods for extending the input dynamic range of hearing aids are described in Chap. 8.

Many hearing aids incorporate two or more microphones and can use these to create directional characteristics. This can be useful when listening in noisy situations provided that the directional characteristic can be "pointed" toward the desired sound source and away from the undesired signal. Chapter 3 describes the directional characteristics of such systems, including the effects of microphone position and mismatched microphones. Chapter 4 describes how more precisely focused directional characteristics can be achieved via signal processing and the transfer of signals between bilaterally fitted hearing aids. Chapters 6 by Souza, 7 by Akeroyd and Whitmer, and 8 consider the effects of directional characteristics on speech perception, spatial hearing, and music perception, respectively.

Chapter 3 also describes transducers that transmit sound by bone conduction; the transducer vibrates the bones of the skull and the vibrations are transmitted through the skull to the cochlea. Such transducers are useful for two types of hearing loss:

1. Permanent conductive (or mixed conductive and sensorineural) hearing loss that is not responsive to surgical or pharmaceutical intervention, where sound is not transmitted effectively to the cochlea by the usual air conduction pathway; the special transducers bypass the conductive mechanism and effectively eliminate the conductive component of the hearing loss.
2. Unilateral hearing loss. For people with unilateral hearing loss, a major problem is a reduced ability to hear sounds coming from the side with hearing loss (Moore and Popelka 2013). To alleviate this problem, sound can be picked up by a microphone on the side with hearing loss and transmitted to the opposite functioning cochlea via bone conduction.

In either case, the sound may be delivered via a vibratory transducer mounted on a pedestal that penetrates the skin and is implanted in the skull, via an active or passive device fixed to the skull underneath the intact skin, or via a dental device held against the teeth.

1.3.6 Signal Processing in Hearing Aids

Chapter 4 describes the signal processing that is used in digital hearing aids. Such processing has become much more elaborate and more complex over successive generations of digital hearing aids. The signal-processing algorithms can be broadly divided into three categories: (1) processing to restore audibility, including multichannel amplitude compression and frequency lowering; (2) “sound cleaning,” via partial removal of background noise, reduction of acoustic feedback (whistling), reduction of wind noise, and selective attenuation of intense transient sounds; and (3) automatic environment classification to allow the hearing aid to change its settings automatically and appropriately in different listening situations.

Most of the signal processing in hearing aids operates in a frequency-dependent manner. To achieve this, a running spectral analysis (a time-frequency analysis) of the input signal is required. Chapter 4 gives an overview of the different types of time-frequency analysis that are used in hearing aids and discusses the advantages and disadvantages of each approach. The chapter then describes the multichannel amplitude compression that is used almost universally in hearing aids to compensate for the reduced dynamic range of hearing-impaired people (as described in Chap. 6). The main aim of multichannel compression is to restore the audibility of weak sounds while preventing intense sounds from becoming uncomfortably loud. Compression systems vary in how rapidly they react to changes in the input sound level, and they can be broadly classified as fast acting or slow acting. The compression speed varies markedly across different brands of hearing aid. The advantages and disadvantages of fast-acting and slow-acting amplitude compression are discussed.

When a person has a severe high-frequency loss, it may be difficult to apply sufficient gain (amplification) to restore audibility. A potential solution to this problem is to apply frequency lowering. With this, high frequencies are moved downward to a frequency range where the hearing loss is less severe. Chapter 4 describes the various methods that are used in hearing aids to implement frequency lowering. The chapter also describes the outcomes of studies evaluating the benefits of frequency lowering. Although the benefits of frequency lowering for speech intelligibility in everyday life are still uncertain, frequency lowering usually does not lead to poorer intelligibility, and it has some beneficial side effects. Specifically, frequency lowering reduces the likelihood of acoustic feedback and, because lower gains are required to restore audibility, it also reduces the possibility of damage to residual hearing.

Chapter 4 next considers various ways of “cleaning” sounds with the goals of improving the intelligibility of speech in noisy situations and improving sound quality and listening comfort. Although most hearing aids incorporate some form of noise reduction, this mainly improves listening comfort rather than intelligibility. Substantial gains in intelligibility can be achieved by the use of directional microphones (see also Chaps. 6 and 7) but there are still limitations. Other forms of processing include simulation of the effects of the pinna (for BTE aids), partial compensation for reverberation, wind noise reduction, and suppression or cancellation of acoustic feedback. Although hearing aids can improve the ability to understand speech in noise, the performance of hearing-impaired people wearing hearing aids rarely approaches that of normal-hearing people. It is clear that much work remains to be done in this area.

The last part of Chap. 4 deals with the automatic classification of environments. Many hearing aids incorporate methods for classifying acoustic environments, for example, speech in quiet, speech in noise, or music. Once an environment has been identified, the settings of a hearing aid may be adjusted automatically. For example, if speech in noise is detected, a highly directional microphone characteristic might be selected, whereas if music is detected, an omnidirectional microphone might be selected.

1.3.7 Wireless Connectivity and Power Requirements

Chapter 5 describes ways in which signals can be transmitted wirelessly from various external devices (remote microphones, conventional and mobile telephones, televisions, stereos, computers, tablets) to hearing aids and between bilaterally fitted hearing aids. Transmission from remote devices can provide an improvement in signal-to-noise ratio and can largely eliminate adverse effects of room echoes and reverberation. This in turn leads to improved sound quality and a better ability to understand speech, especially in adverse listening conditions. Improvements in signal-to-noise ratio and sound quality can also be achieved with wireless transmission directly to hearing aids from a variety of public address systems such as in theaters, lecture halls, places of worship, airports, train stations, and bus stations. The variety of methods used for wireless transmission to hearing aids and the benefits and limitations of each method are described. Factors affecting usability are discussed, including battery consumption. The problem of a lack of international standardization is also discussed.

1.4 Perception of Sound via Hearing Aids

1.4.1 Speech Perception

Chapter 6 describes the effects of hearing loss and of hearing aids on the perception of speech. Sensorineural hearing loss nearly always leads to problems in understanding speech, especially in noisy or reverberant situations. Generally, the greater the hearing loss, as measured using the audiogram, the greater the difficulty in understanding speech. However, there is often considerable variability in speech perception ability across people with similar audiograms, and the factors underlying this variability are poorly understood.

In cases of mild to moderate hearing loss, hearing aids can provide substantial benefit. For severe sensorineural hearing loss, the ability to analyze and discriminate sounds that are well above the detection threshold is reduced and this limits the benefit provided by hearing aids. In cases of profound or total loss of hearing, so little auditory function remains that a conventional hearing aid is ineffective and a cochlear implant may be a better alternative. Cochlear implants, which are widely used to treat severe and profound hearing loss in both children and adults, bypass the missing or remaining dysfunctional sensory structures altogether and activate the remaining VIIIth cranial nerve fibers directly; see Clark et al. (1987) and Zeng et al. (2003). For people with residual low-frequency hearing but severe or profound hearing loss at medium and high frequencies, the combination of a hearing aid and a cochlear implant may be more effective than either alone (Dorman and Gifford 2010; Zhang et al. 2014).

Chapter 6 considers the influence of several factors that may contribute to individual differences in auditory performance and benefit from hearing aids, including individual differences in the underlying patterns of cochlear and neural damage and individual differences in cognitive ability. The chapter describes how sensorineural hearing loss can arise from many causes, including damage to (1) the outer hair cells, affecting the operation of the active mechanism in the cochlea and leading to loss of sensitivity to weak sounds and reduced frequency selectivity and loudness recruitment (Robles and Ruggero 2001); (2) the inner hair cells, leading to more “noisy” transmission of information and, in extreme cases, dead regions in the cochlea (Moore 2001); (3) the stria vascularis, disturbing the metabolism of the cochlea and affecting the operation of both inner and outer hair cells and neural processes (Schmiedt 1996); (4) the synapses between inner hair cells and neurons, with effects similar to those of inner hair cell damage (Kujawa and Liberman 2009); and (5) neurons in the auditory nerve and higher up in the auditory pathway, causing a general reduction in the ability to discriminate sounds (Schuknecht 1993) and in the ability to combine input from the two ears (Durlach et al. 1981; Moore et al. 2012a).

Some speech perception problems of the hearing impaired, such as a poor ability to understand soft speech, arise primarily from reduced audibility; part of the speech spectrum falls below the elevated absolute thresholds (Humes and Roberts 1990). These problems can, in principle, be alleviated via the amplification provided by hearing aids. However, hearing aids have proved to be much less effective in improving the ability to understand speech in noisy or reverberant conditions (Plomp 1978). Chapter 6 reviews the effects of hearing loss on speech perception and discusses the extent to which hearing aids can compensate for those effects. The chapter describes the acoustic cues that are present in the speech signal and how those cues are affected by the signal processing that is used in hearing aids, especially multichannel amplitude compression. The chapter also describes several different types of masking that can affect speech perception and describes how hearing aids affect these different types of masking. The types include (1) “energetic masking,” which occurs when the pattern of responses in the auditory nerve is similar for the masker alone and the masker plus the target speech (Brungart et al. 2006); (2) “modulation masking,” which occurs when the patterns of amplitude modulation in the masker make it more difficult to detect and discriminate patterns of amplitude modulation in the target speech (Stone et al. 2011; Jørgensen et al. 2013); and (3) “informational masking,” which occurs when the target speech and masker are confusable and/or similar, for example, when two people with similar voices speak at the same time (Brungart et al. 2001; Lunner et al. 2012).

The majority of hearing aid users are older than 60 years of age (see Chap. 2), and some auditory abilities, such as sensitivity to temporal fine structure, decline with increasing age even for people whose audiograms remain within the normal range (Moore et al. 2012a, b; Füllgrabe et al. 2015). Reduced sensitivity to temporal fine structure seems to be associated with difficulty in understanding speech in background sounds (Füllgrabe et al. 2015). It may also be associated with a reduced ability to judge harmony in music (Bones and Plack 2015). In addition, cognitive abilities tend to decline with increasing age, and this probably also contributes to difficulty in understanding speech in background sounds (Akeroyd 2008; Füllgrabe et al. 2015). At present, hearing aids do little to alleviate the effects of suprathreshold deficits in auditory or cognitive processing, except indirectly, for example, by the use of directional microphones to improve the signal-to-background ratio.

Chapter 6 includes a review of the effectiveness for speech perception of the major types of signal processing that are used in hearing aids. An important take-home message is that the effectiveness of different types of processing varies across acoustical situations and across individuals. Hence, when programming a hearing aid and selecting the types of signal processing that should be activated for optimal speech understanding, every individual's hearing ability should be treated as unique. Much remains to be learned as to how to set up a hearing aid optimally for the individual listener.

1.4.2 Spatial Perception

Chapter 7 describes the influence of hearing loss and hearing aids on spatial perception. Spatial perception refers to the ability to use spatial cues to better understand speech in the presence of background noise or to judge the location of sounds in space. Generally, the accuracy of sound localization is greatest for judgments of azimuth (along the left–right dimension in the horizontal plane). Accuracy is somewhat poorer for front–back and up–down sound localization. The chapter considers each of these separately.

Chapter 7 describes the two main methods by which the accuracy of sound localization has been measured. In one method, called "source discrimination," two sounds are presented from different directions, ϕ_1 and ϕ_2, and the listener has to decide if the order of presentation was ϕ_1 then ϕ_2 or ϕ_2 then ϕ_1. Usually, a threshold corresponding to the smallest difference between ϕ_1 and ϕ_2 that can be reliably detected is determined. This is usually called the minimum audible angle (MAA) (Mills 1958), although in Chap. 7 it is called the $\mathrm{JND}_{\mathrm{MAA}}$ to emphasize that it is a just-noticeable difference. In the second method, called "source identification," a sound is presented from any one of an array of loudspeakers (or from one of an array of "virtual" sources using sounds presented via headphones). The task is to report the direction of the sound, for example, by identifying the loudspeaker that it emanated from or by pointing toward the perceived source. Several different measures have been used to quantify localization accuracy using the source identification method, and these measures and their interrelations are described in Chap. 7; the recommended measure is the root-mean-square error in degrees, *D*. To allow comparison of results across studies using different measures, in Chap. 7 all measures are converted to *D* where possible.

Chapter 7 presents evidence showing that for left–right sound localization, hearing-impaired people tend to be less accurate than normal-hearing people. However, some of the effects that have been attributed to hearing loss may be a consequence of age because the experiments have often compared results for young normal-hearing and older hearing-impaired subjects; see Moore et al. (2012b) and Füllgrabe et al. (2015) for discussion of this point. Bilaterally fitted hearing aids on average lead to a slight worsening in left–right discrimination relative to unaided listening, although Akeroyd and Whitmer argue that this effect is too small to have a noticeable effect in everyday life. However, unilateral aiding can lead to a distinct worsening in left–right localization. Hence, aided performance is almost always better with bilateral than with unilateral hearing aids.

Hearing impairment tends to have a greater deleterious effect on sound localization in the front–back and up–down dimensions than on sound localization in the left–right dimension. Hearing aids do not generally improve front–back or up–down discrimination, but they also usually do not make it markedly worse.

1.4.3 Music Perception

Chapter 8 describes the effect of hearing aids on the perception of music. Most hearing aids are designed primarily with the goal of improving the ability to understand speech, but many users of hearing aids also want to listen to music via their aids. Music is a more difficult signal to handle than speech because its spectrum varies widely across different types of music and different listening venues, it can cover a very wide range of sound levels, it can have a high crest factor (ratio of peak-to-root-mean-square value), and there can be significant energy in music over a very wide range of frequencies (Chasin and Hockley 2014). Unlike speech, there is no such thing as a “typical” music spectrum.

Chapter 8 considers the characteristics required of microphones, amplifiers, signal processors, and receivers to provide a relatively undistorted, high-fidelity, noise-free representation of musical signals. In many current hearing aids, the requirements are not met, so distortion can occur with high-level inputs and sound quality can sometimes be low (Madsen and Moore 2014). Chapter 8 describes ways in which these problems can be alleviated. The chapter also describes the effects on music perception of some of the signal-processing algorithms that are used in hearing aids, including wide dynamic range compression (WDRC), active sound feedback cancellation, frequency lowering, wind noise reduction, other types of noise reduction, and acoustic transient reduction. WDRC can help to make soft passages in music audible while preventing loudness discomfort for high-level sounds, but there are side effects of WDRC processing that can degrade sound quality and make it harder to hear out individual instruments or voices in a mixture (Madsen et al. 2015). An alternative approach called adaptive dynamic range optimization (ADRO) is described in Chap. 8. This is basically a slow-acting form of multichannel amplitude compression that may be preferable to fast-acting compression for listening to music.

Chapter 8 also considers possible deleterious effects of the time delays produced by the signal processing in hearing aids (including frequency-dependent delays), of the limited bandwidth of hearing aids, and of ripples in the frequency response of hearing aids. A general take-home message in relation to the use of hearing aids for music listening is “do no harm.” In other words, perform the minimum signal processing necessary to make the music audible and comfortably loud.

1.5 Clinical Verification of Hearing Aid Performance

Chapter 9, by Munro and Mueller, is concerned mainly with methods for fitting hearing aids and methods for verifying that the fitting is correct in terms of the sound delivered to the tympanic membrane. It is widely accepted that hearing aids should be fitted according a method that has been validated in clinical studies. Such methods include Desired Sensation Level (DSL) v5 (Scollie et al. 2005), National Acoustics Laboratories, Non-Linear, version 1 (NAL-NL1; Byrne et al. 2001) and version 2 (NAL-NL2, Keidser et al. 2011), and CAM2 (previously called CAMEQ2-HF; Moore et al. 2010). As stated by Munro and Mueller, “no validated prescription method has been clearly shown to be superior to any of the other methods in terms of patient benefit (e.g., greater satisfaction, less residual disability). However, clinical studies have clearly shown that when a well-researched prescriptive approach is used and appropriate gain is delivered to the tympanic membrane across frequencies, speech intelligibility is enhanced, and there is improved patient benefit and satisfaction.” Despite this, many manufacturers of hearing aids use their own proprietary fitting methods that have not been subjected to independent peer-reviewed evaluation. Even when the manufacturer’s fitting software allows a “standard” prescriptive method to be selected, the real-ear output of the hearing aid often differs from that prescribed by the selected method (Aazh et al. 2012). Hence, it is important that the output of the hearing aid is checked and adjusted where necessary using measurements with a probe microphone placed in the ear canal close to the tympanic membrane.

There are many types of test signals that can be used for verification of hearing aids, including pure tones, swept tones, narrow bands of noise, speech, and artificial speech like signals (Dreschler et al. 2001; Holube et al. 2010). Generally, speech or speech like signals are preferred, because other signals may activate signal-processing algorithms such as acoustic feedback cancellation or noise reduction, giving misleading results. There are also many ways of expressing the output of the hearing aid in the ear canal, including insertion gain, real-ear aided gain, and output level in decibels SPL. Chapter 9 considers these various measures in detail. Munro and Mueller advocate the use of output level in decibels SPL because “it enables easy visualization of the interrelationship between assessment data, the level of unamplified speech, and the amplification characteristics, which are typically measured in different units and at different reference points.” The authors also advocate use of the speech intelligibility index (ANSI 1997) as a standard way of assessing the extent to which the audibility of speech has been restored, although they point out some limitations of this approach.

Chapter 9 describes many of the technical and practical problems encountered when making real-ear measurements of the output of hearing aids. Methods of solving or alleviating those problems are described. The chapter also describes methods for verifying the correct operation of features of the hearing aid such as frequency lowering, directional microphones, noise reduction, and acoustic feedback reduction.

1.6 Validation of Hearing Aid Performance

Chapter 10, by Whitmer, Wright-Whyte, Holman, and Akeroyd, focuses on validation of the performance of hearing aids, especially via the use of questionnaires. The main goal is to assess whether the hearing aids meet the needs of the individual user, in other words to assess “if they alleviate disability and handicap, or if they relieve restrictions or limitations due to hearing loss.” As well as being used for assessing the benefit of hearing aids relative to listening unaided for a specific individual, questionnaires may be used to evaluate the effectiveness of different types of signal processing (e.g., linear amplification vs. WDRC; Moore et al. 1992) or different types of hearing aids (e.g., bone-anchored vs. tooth-mounted aids for unilateral hearing loss; Moore and Popelka 2013).

Chapter 10 reviews the very large number of questionnaires that have been developed for the purpose of validation of hearing aid performance and describes the great variety of domains that have been addressed. The domains include frequency of use, effectiveness for speech in quiet, effectiveness for speech in noise and/or reverberation, ability to use the telephone, sound quality, extent of auditory fatigue, ease of listening, changes in self-esteem, ease of operation, aversiveness of sounds, loudness of sounds, binaural and spatial hearing, effects on social life, quality of life, residual activity limitation, and willingness to purchase. Some questionnaires focus on specific populations, such as users of bone-anchored hearing aids (see Chap. 3), or on specific problems, such as the quality of the aid user’s own voice.

Chapter 10 discusses a problem in the use of questionnaires to evaluate the benefit provided by hearing aids. A respondent may be unable to judge the benefit in a specific situation and may give a response reflecting the difficulty experienced in that situation rather than reflecting the benefit. This problem can be reduced by asking respondents to describe situations that they have experienced and can comprehend and then asking them to rate difficulty without and with hearing aids only in the experienced situations. Even then, the task may be difficult because the respondents usually have not directly compared experiences without and with hearing aids. Assuming that they are currently using hearing aids, they have to imagine and remember their listening experience before they were fitted with hearing aids, and this process of recall may be influenced by many biases.

Another issue addressed in Chap. 10 is "rational noncompliance." This occurs when a client rejects an option, such as having two aids as opposed to one, where there would be objective benefit from having two aids, at least in terms of speech intelligibility and sound localization. The key idea here is that of "burdens." For example, an extra hearing aid requires more batteries and more maintenance and may involve greater perceived stigma. The net benefit of an extra hearing aid reflects a balance between performance improvements and burdens. Put simply, an extra hearing aid may not be worth the bother.

Chapter 10 also discusses the distinction between benefit and satisfaction. If a person has unrealistically high expectations of hearing aids, then even if the hearing aids lead to measurable improvements in speech perception, satisfaction will be low. In contrast, if expectations are low, even small objective benefits may be associated with high satisfaction.

Finally, Chap. 10 discusses the use of tests of speech perception for validation of hearing aids. Although many speech tests have been developed and used in clinical evaluations of different types of hearing aids or different types of signal processing within hearing aids, speech tests are rarely used to evaluate the effectiveness of hearing aids for individual clients in the clinic. A basic problem is that the majority of speech tests are artificial, involving listening without visual information to isolated words or sentences without context and in the absence of reverberation. The results of such tests may give little insight into the effectiveness of hearing aids in everyday life. Although some speech tests with greater realism have been developed, these mostly focus on one aspect of realism, for example, providing context, simulating the effects of reverberation, or providing visual information. A speech test that simulates realistic listening conditions and that could be used for validation in the clinic has yet to be developed.

1.7 Concluding Remarks

This chapter has given an overview of the many factors that are relevant to the design, fitting, and usefulness of hearing aids. It has also indicated areas of uncertainty, where further research is needed. Chapter 11, by Gerald Popelka and Brian Moore, describes possible future directions for hearing aids and hearing aid research.

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Gerald Popelka declares that he has no conflict of interest.

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Chapter 2
Epidemiology of Hearing Impairment

Gary Curhan and Sharon Curhan

Abstract Hearing impairment is the most prevalent sensory deficit, affecting approximately 30 million (12.7 %) individuals in the United States in both ears and 48 million (20.3 %) individuals in the United States in at least one ear. Nevertheless, NIH estimates suggest that only 20 % of people who could potentially benefit from a hearing aid seek intervention. Globally, approximately 5.3 % of the world's population, or 360 million individuals, suffer from hearing impairment that is considered to be disabling by WHO standards. Hearing impairment is a condition that can develop across the life span, and the relations between specific risk factors and hearing impairment may vary with age. The etiology of hearing impairment is complex and multifactorial, representing the cumulative influences of an amalgam of factors, such as aging, genetic, epigenetic, environmental, health comorbidity, diet and lifestyle factors, as well as the complex potential interactions among these factors, that may all contribute to its development. Identification of risk factors for hearing impairment may provide us with a better understanding of the cellular and molecular mechanisms associated with acquired hearing impairment and could aid efforts toward prevention, early detection, and delay of progression. This chapter provides an overview of the epidemiology of hearing impairment in the United States and worldwide, including information on incidence, prevalence, and a discussion of risk factors that have been identified as potential contributors.

Keywords Adults • Diet • Environment • Genetics • Global • Incidence • Lifestyle • Medications • Noise • Pediatric • Prevalence • Risk factors

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2.1 Introduction

Hearing impairment is the most prevalent sensory deficit, affecting approximately 30 million (13 %) individuals in the United States in both ears and 48 million (20 %) individuals in the United States in at least one ear (Lin et al. 2011a). Globally, approximately 5.3 % of the world's population, or 360 million individuals, suffer from hearing impairment that is considered to be disabling by World Health Organization (WHO) standards (Informal Working Group 1991; WHO 2012). The individual and societal burdens imposed by hearing impairment are considerable. Hearing impairment has been found to be associated with poorer quality of life, increased comorbidities, difficulties with functional activities, lower work productivity, and reduced income (Peters et al. 1988; Bess et al. 1989; Uhlmann et al. 1989; Mulrow et al. 1990; Carabellese et al. 1993; Campbell et al. 1999; Crews and Campbell 2004; Kochkin 2007). Data from the National Health Interview Survey (2006) show that one in six American adults reports trouble hearing (Pleis and Lethbridge-Cejku 2007). Estimates of the overall prevalence of audiometrically measured hearing impairment in the United States from a nationally representative sample, the National Health and Nutrition Examination Survey (NHANES), suggest that for individuals aged 12 years and older, nearly one in eight has bilateral hearing impairment and almost one in five has either unilateral or bilateral hearing impairment, with prevalence of any hearing impairment increasing considerably with each decade of age (Pleis and Lethbridge-Cejku 2007; Lin et al. 2011a).

Identifying and quantifying the proportion of US individuals who would likely benefit from the use of hearing aids could inform healthcare policy and aid public health efforts to improve the accessibility, affordability, and outcome of hearing healthcare. In general, individuals who have a mild-to-moderate bilateral hearing impairment and are experiencing communication difficulties are those most likely to benefit. Based on the criterion established by the Federal Interagency Working Group for Healthy People 2010, the population most likely to benefit includes those US adults with air conduction audiometric thresholds in the worse ear pure-tone average (PTA) from 1 to 4 kHz, $PTA_{(1,2,3,4\ kHz)} \geq 35$ dB hearing level (HL), and excludes the approximately 10 % of those with reversible conductive hearing impairment or who have profound losses and would be candidates for cochlear implants. Data from NHANES and the National Health Interview Survey (NHIS) indicate that of US adults aged 20–69 years who could potentially benefit from hearing aid use, only 16 % have ever used them. For US adults aged 70 years and older, it is estimated that only 29 % of those who could potentially benefit have ever used hearing aids (National Institute on Deafness and Other Communication Disorders [NIDCD] 2012). Although the number of US individuals who are most likely to benefit from hearing aid use is not available in the literature, in the United Kingdom it is reported that more than six million adults could benefit from hearing aids (Loss 2014). On average, hearing aid users have lived with their hearing impairment for 10 years before seeking intervention. Factors that influence hearing aid use include cost, stigma, perceived versus actual benefit, and accessibility to hearing healthcare (NIDCD 2009).

An amalgam of factors, such as aging, genetic, epigenetic, environmental, health comorbidity, diet, and lifestyle, as well as the complex potential interactions among these factors, contribute to the development of hearing impairment. Identification of risk factors for hearing impairment may provide a better understanding of the cellular and molecular mechanisms associated with acquired hearing impairment and could aid efforts toward prevention, early detection, and delay of progression.

2.2 Definitions

The WHO classification of hearing impairment is based on the PTA hearing threshold levels at four frequencies (0.5, 1, 2, and 4 kHz) in the individual's better hearing ear. The classifications range from "no impairment" to "profound impairment." Disabling hearing impairment in adults is defined as a permanent unaided better ear $PTA_{(0.5,1,2,4\ kHz)}$ >40 dB HL and in children aged younger than 15 years as >30 dB HL. Numerous alternative definitions have been used in both clinical and research settings. Thus, comparisons among reports can be challenging. For example, hearing impairment has been defined according to better ear hearing level, worse ear hearing level, or average hearing level across ears; as high-frequency, speech-frequency range, low-frequency, or any type of impairment; and as varying PTA threshold cutoff points (WHO).

A systematic analysis of the data on the epidemiology of hearing impairment in newborns, children, and adolescents in the United States faces challenges similar to those for adults with respect to inconsistent methodology and definitions of hearing impairment across studies. The epidemiology of pediatric hearing impairment is discussed in Sect. 2.10.

2.3 Prevalence

Prevalence refers to the proportion of a population with a condition, such as hearing impairment, at a given point in time. A valuable resource for deriving prevalence estimates in the US population is the NHANES, which provides nationally representative data from an ongoing survey of the US population. Data collected from the 2001 through 2008 NHANES provided estimates of the overall prevalence of hearing impairment among all US individuals aged 12 and older, using the WHO definition of hearing impairment: $PTA_{(0.5,1,2,4\ kHz)}$ >25 dB HL in the better ear. There were 30 million individuals, comprising 13 % of the US population, who had bilateral hearing impairment. In addition, 48.1 million Americans (20 %) had unilateral or bilateral hearing impairment (see Table 2.1). The prevalence of hearing impairment increased with increasing decade of age, was higher for men than for women, and was higher for white than for black individuals in most age categories (Lin et al. 2011a).

Table 2.1 Prevalence of hearing loss[a] [≥25 dB HL (bilateral and unilateral)] in the United States by age, sex, race, and ethnicity[b]

g									
Variable	Female	Male	White	Black	Hispanic	Overall prevalence	No. with hearing loss (in millions)	Overall prevalence	No. with hearing loss (in millions)
Age, years									
12–19	0.42 (0–0.91)	0.20 (0–0.41)	0.26 (0–0.66)	0.48 (0.11–0.85)	0.43 (0.04–0.82)	0.31 (0.04–0.57)	0.10	2.3 (1.5–3.1)	0.76
20–29	0.35 (0–0.79)	0.48 (0–1.4)	0.43 (0–1.3)	0.63 (0–1.9)	0.35 (0–0.90)	0.42 (0–0.97)	0.16	3.2 (1.4–5.1)	1.2
30–39	0.79 (0–1.8)	2.5 (0.14–4.9)	1.8 (0–3.8)	1.7 (0–3.9)	1.6 (0.22–3.1)	1.6 (0.23–3.1)	0.68	5.4 (3.3–7.6)	2.3
40–49	4.5 (0.94–8.1)	8.7 (5.0–12.4)	7.4 (4.5–10.3)	1.3 (0–3.3)	7.3 (2.0–12.5)	6.5 (4.1–8.8)	2.8	12.9 (9.8–15.9)	5.6
50–59	6.1 (3.6–8.6)	20.3 (14.5–26.2)	14.5 (9.9–19.2)	7.1 (3.0–11.2)	13.8 (6.4–21.2)	13.1 (9.4–16.8)	4.4	28.5 (23.3–33.7)	9.6
60–69	16.8 (12.1–21.5)	39.2 (31.7–46.8)	26.6 (21.1–32.1)	15.9 (9.8–22.1)	28.9 (17.0–40.8)	26.8 (22.3–31.4)	5.7	44.9 (40.9–48.9)	9.5
70–79	48.5 (38.5–58.5)	63.4 (56.2–70.5)	55.8 (47.6–63.9)	39.0 (26.2–51.7)	66.8 (52.3–81.2)	55.1 (48.0–62.2)	8.8	68.1 (61.2–75.1)	10.8
≥80	75.6 (69.7–81.5)	84.6 (79.0–90.3)	81.5 (78.5–84.5)	54.8 (40.6–69.0)	60.7 (34.8–86.6)	79.1 (76.0–82.2)	7.3	89.1 (86.1–92.0)	8.3
Estimated total no. of individuals with hearing loss (in millions)						30.0		48.1	

Data from NHANES 2001–2008

Adapted from Table 2.1 in Lin, F. R., Niparko, J. K., & Ferrucci, L. (2011). Hearing loss prevalence in the United States. *Archives of Internal Medicine*, 171(20), 1851–1852

[a] Hearing loss is defined based on the pure-tone average of thresholds at 500, 1,000, 2,000, and 4,000 Hz

[b] The prevalence values are percentages (CI in parentheses) except where noted

Numerous methods have been used to track the prevalence of hearing impairment over time, and estimates have varied. The “gold standard” for hearing assessment is generally considered to be conventional pure-tone audiometry. However, owing to the high cost and logistic limitations associated with conducting audiometric testing in large populations, only a few such studies of nationally representative samples have been performed. Frequently, assessment of self-reported hearing status has been used to examine the prevalence of hearing impairment in large nationally representative interview surveys in the United States. Unfortunately, there are differences among studies in the wording of questions. Thus, estimates of prevalence based on self-report should be interpreted cautiously. Nevertheless, estimates based on the most recent studies in the United States and globally provide compelling findings and strongly suggest that a substantial proportion of the world’s population can be characterized as suffering from hearing impairment to some degree.

Given the importance of tracking the prevalence of hearing impairment over time as a basis for effective formulation and evaluation of public health policy related to hearing health, strategies have been developed for examining population trends in US adults (Ikeda et al. 2009). The prevalence of bilateral hearing impairment in the speech-frequency range (defined as $PTA_{(0.5,1,2,4\,kHz)}$ >25 dB HL) declined in the 1990s and stabilized in the early 2000s. For example, the prevalence of bilateral hearing impairment in men was 9.6 % [95 % confidence intervals (CIs): 7.7, 11.8 %] in 1978, rose to 12.2 % (95 % CI: 10.1, 14.7 %) in 1993, declined to 8.1 % (95 % CI: 7.0, 9.5 %) in 2000, and then remained somewhat stable (8–9 %) until 2006. The age-standardized prevalence of bilateral hearing impairment among women during this period was lower than for men, yet followed a similar pattern.

Although some reports indicate that the overall risk of hearing impairment may have decreased over time (Hoffman et al. 2010; Zhan et al. 2010), the number of individuals with hearing impairment is expected to increase as a result of the aging of the US population. The finding of generational differences in age at onset of hearing impairment (Zhan et al. 2010) underscores the importance of identifying potentially modifiable risk factors.

2.3.1 Prevalence of Hearing Impairment in US Adolescents

Between the 1988–1994 and 2005–2006 NHANES evaluations, the prevalence of any hearing impairment in US adolescents aged 12–19 years increased significantly from 14.9 % (CI: 13.0–16.9 %) to 19.5 % (CI: 15.2–23.8 %). This translates to approximately 6.5 million US adolescents with hearing impairment in 2005–2006, a 31 % increase (Shargorodsky et al. 2010a). In 2005–2006, hearing impairment was more commonly unilateral, with a prevalence of 14.0 %, and high frequency, with a prevalence of 16.4 % (see Table 2.2). In this age group, the prevalence of hearing impairment did not differ by age or race/ethnicity. However, females were 24 % less likely than males to have any hearing impairment. The reason for the increase in prevalence over two decades is not clear.

Table 2.2 Prevalence of hearing loss in United States adolescents in two NHANES cycles

	No. (prevalence, %) [95 % CI] by hearing threshold (HL severity)		
	>15 dB HL (slight or worse)	>15 to <25 dB HL (slight)	≥25 dB HL (mild or worse)
NHANES III (1988–1994)			
Any HL[a]	480 (14.9) [13.0–16.9]	360 (11.4) [9.7–13.1]	120 (3.5) [2.5–4.5]
Any high-frequency HL	423 (12.8) [11.1–14.5]	339 (10.1) [8.5–11.6]	84 (2.7) [1.7–3.7]
Any low-frequency HL	186 (6.1) [4.5–7.6]	151 (5.2) [3.9–6.5]	35 (0.9) [0.1–1.7]
Unilateral HL	335 (11.1) [9.5–12.8]	278 (9.3) [7.9–10.7]	57 (1.8) [0.9–2.8]
Unilateral high-frequency HL	304 (9.6) [8.1–11.2]	245 (7.7) [6.4–9.1]	59 (1.9) [0.9–2.8]
Unilateral low-frequency HL	140 (5.0) [3.4–6.4]	113 (4.3) [3.0–5.5]	27 (0.7) [0.0–1.4]
Bilateral HL	145 (3.8) [2.6–4.9]	120 (2.9) [2.0–3.8]	25 (0.8) [0.3–1.4]
Bilateral high-frequency HL	119 (3.2) [2.2–4.1]	94 (2.3) [1.6–3.0]	25 (0.8) [0.3–1.4]
Bilateral low-frequency HL	46 (1.1) [0.6–1.7]	38 (0.9) [0.4–1.4]	8 (0.2) [0.0–0.5]
NHANES 2005–2006			
Any HL[a]	333 (19.5) [15.2–23.8]	239 (14.2) [10.6–17.8]	94 (5.3) [3.6–6.9]
Any high-frequency HL	279 (16.4) [13.2–19.7]	219 (11.7) [9.4–14.1]	60 (4.7) [3.3–6.1]
Any low-frequency HL	155 (9.0)] [5.6–12.5]	126 (6.5) [3.5–9.4]	29 (2.5) [1.4–3.7]
Unilateral HL	234 (14.0) [10.4–17.6]	191 (11.3) [8.2–14.5]	43 (2.7) [1.4–3.9]
Unilateral high-frequency HL	209 (12.6) [9.9–15.3]	167 (9.8) [7.8–11.8]	42 (2.8) [1.7–3.9]
Unilateral low-frequency HL	113 (6.8) [3.8–9.8]	90 (5.3) [2.7–8.0]	23 (1.5) [0.6–2.3]
Bilateral HL	99 (5.5) [3.9–7.1]	80 (4.7) [3.5–5.8]	19 (0.8) [0.1–1.5]
Bilateral high-frequency HL	70 (3.8) [2.5–5.1]	52 (3.0) [2.1–3.9]	18 (0.8) [0.1–1.5]
Bilateral low-frequency HL	42 (2.2) [1.5–3.0]	36 (2.0) [1.4–2.7]	6 (0.2) [0.0–0.5]

Adapted from Table 2.2 in Shargorodsky, J., Curhan, S. G., Curhan, G. C., & Eavey, R. (2010). Change in prevalence of hearing loss in US adolescents. *JAMA*, 304(7), 772–728

[a]Any hearing loss refers to unilateral or bilateral at low or high frequencies. Data from NHANES 1988–1994 and 2005–2006. Low-frequency hearing loss is defined based on the pure-tone average of thresholds at 500, 1,000, and 2,000 Hz; high-frequency hearing loss is defined based on the pure-tone average of thresholds at 3,000, 4,000, 6,000, and 8,000 Hz

2.3.2 Prevalence of Hearing Impairment in US Adults

NHANES data from 2003 to 2004 show that the overall prevalence of unilateral hearing impairment (defined as $PTA_{(0.5,1,2,4\ kHz)}$ >25 dB HL in one ear only) and bilateral speech-frequency hearing impairment (defined as $PTA_{(0.5,1,2,4\ kHz)}$ >25 dB HL in both ears) among the US population aged 20–69 years was 16.1 % (7.3 % bilateral and 8.9 % unilateral hearing impairment). There were notable differences in the prevalence of hearing impairment according to various demographic factors. Speech-frequency and high-frequency hearing impairment were more prevalent for males and older individuals and more prevalent for white than black individuals. Among Mexican Americans, the prevalence was lower than for whites in some age groups and higher in others. Hearing impairment was more prevalent among those with less education, who reported occupational or leisure-time noise exposure, or firearm use, and those with hypertension, with diabetes mellitus, and who reported heavy smoking (>20 pack-years) (Agrawal et al. 2008).

2.3.3 Prevalence of Hearing Impairment in the Older US Population

In a 2006 population-based study of individuals older than 70 years, the overall prevalence of hearing impairment, defined according the WHO standard of speech-frequency $PTA_{(0.5,1,2,4\ kHz)}$ in both ears >25 dB, was 63.1 %. More than 80 % of individuals older than 85 years had hearing impairment. In this older population, the odds of hearing impairment were higher among men [odds ratio (OR) = 1.67] and lower for blacks than for whites (OR = 0.32). However, no association between history of noise exposure or other medical conditions, such as diabetes, smoking, hypertension, or stroke, was observed (Lin et al. 2011a). Prevalence estimates in older populations have differed among studies, possibly a reflection of differences in definitions used and/or the demographic characteristics of the study populations examined. Among participants older than age 70 in the Beaver Dam, Wisconsin, Epidemiology of Hearing Impairment Study (EHLS), the prevalence of speech-frequency $PTA_{(0.5,1,2,4\ kHz)}$ >25 dB HL in the worse ear was 73 % (Cruickshanks et al. 1998). Among participants aged 73–84 years in the Health, Aging and Body Composition (Health ABC) Study, the prevalence of $PTA_{(0.5,1,2\ kHz)}$ >25 dB HL in the worse ear was 60 % (Helzner et al. 2005), and among participants older than age 60 in the Framingham Heart Study, the prevalence of $PTA_{(0.5,1,2\ kHz)}$ >25 dB HL in the better ear was 29 % (Gates et al. 1990). Among US adults aged 73–84 years from Pennsylvania and Tennessee who were participants in the Health ABC study, the prevalence of hearing impairment ($PTA_{(0.5,1,2\ kHz)}$ >25 dB HL) was 59.9 % and the prevalence of high-frequency hearing impairment ($PTA_{(2,4,8\ kHz)}$ >40 dB HL) was 76.9 % (Helzner et al. 2005). In an Australian population aged 49 years and older (Blue Mountains Study), the prevalence of any hearing impairment, defined as $PTA_{(0.5,1,2,4\ kHz)}$ >25 dB HL in the better ear, was 33.0 % and age- and sex-specific prevalence rates were comparable to those found in NHANES (Gopinath et al. 2009).

2.4 Incidence

Incidence is the number of cases of a condition that develop during a specified time period. Because the exact time of onset of hearing impairment is often uncertain, the time of diagnosis or onset of symptoms is often used. For this reason, it can be a challenge to compare two or more incidence rates if the same criteria have not been used.

Few longitudinal studies have evaluated the incidence of hearing impairment in the US population. Most studies have been based on restricted populations with regard to age, demographic, or geographic parameters. For example, the 5-year incidence and progression of hearing impairment defined as $PTA_{(0.5,1,2,4\ kHz)}$ >25 dB HL in either ear was examined in a cohort of adults aged 48–92 years in the EHLS (Cruickshanks et al. 2003). The overall 5-year incidence of hearing impairment was 21.4 % and it increased with age. Age-specific incidence rates were higher for men than for women among those aged younger than 70 years. However, there was no difference among older individuals.

2.5 Rate of Threshold Change

Many factors associated with higher incidence or prevalence of hearing impairment have also been examined in relation to the rate of decline in hearing sensitivity. In a Dutch cohort followed for 12 years, the overall deterioration rate (better ear $PTA_{(1–4\ kHz)}$) was 7.3 dB per decade (Linssen et al. 2014). The rate per decade was 5.1 dB for adults aged 24–42 years, 7.6 dB for adults aged 43–62 years, and 12.3 dB for adults aged 63–81 years. Poorer hearing thresholds at baseline were associated with a greater rate of change. In addition, increasing age and male sex were associated with a greater rate of change. In the EHLS, more than half of participants with hearing impairment at baseline demonstrated more than a 5-dB increase in $PTA_{(0.5–4\ kHz)}$ over 5 years of follow-up (Cruickshanks et al. 2003). The risk for progression of hearing impairment increased with age but did not vary by sex and was not associated with level of education or type of occupation. A longitudinal study of changes in hearing thresholds in adults over a 10-year period (Wiley et al. 2008) observed worsening of hearing thresholds across all frequencies, and the frequencies at which changes were of greatest magnitude varied with age. Among individuals aged 48–69 years, the greatest magnitude of change was at higher frequencies (3–8 kHz), whereas among individuals aged 80 years and older, the greatest magnitude was at lower frequencies (0.5–2 kHz). Among younger individuals, the rate of threshold change was greater at high frequencies than at lower frequencies and the absolute difference between rates of change at high versus low frequencies decreased with increasing age. The most important predictors of change in thresholds over 10 years in this study were age, sex, and baseline threshold at the same frequency.

2.6 Risk Factors for Acquired Hearing Impairment

Hearing impairment is a condition that can develop across the life span, and the relationships between specific risk factors and hearing impairment may vary with age. For example, genetic factors and prenatal infections are the primary risk factors for congenital hearing impairment. For children, genetic disorders, meningitis, head trauma, and ototoxic medications are important risk factors. For adults, the etiology of hearing impairment is complex and multifactorial, representing the cumulative influences of aging, excessive noise and other environmental exposures, genetics, medical conditions, medications, and lifestyle factors.

As hearing impairment is often insidious in onset, it can be difficult to identify the site of initial injury or to isolate the contribution of individual precipitants. Although the precise mechanisms that underlie many aspects of hearing impairment remain uncertain, much has been learned from both animal and human studies regarding the underlying pathophysiology, which has helped to inform the examination and identification of risk factors for hearing impairment.

A number of epidemiologic studies have examined nonmodifiable and potentially modifiable risk factors for hearing impairment. However, the results have often been inconsistent. Conflicting findings may be due to differences in definitions of hearing impairment, characteristics of the study cohorts, and variability in gene–risk factor interactions. In general, a higher prevalence of hearing impairment has been cross-sectionally associated with male sex (Agrawal et al. 2008; Fransen et al. 2008; Lin et al. 2011b; Nash et al. 2011; Kiely et al. 2012), white race (Agrawal et al. 2008; Lin et al. 2011b), lower household income level (Lin et al. 2011b), lower educational attainment (Agrawal et al. 2008; Lin et al. 2011b), history of cerebrovascular disease (Kiely et al. 2012), diabetes (Kiely et al. 2012), inflammatory bowel disease (Akbayir et al. 2005), rheumatoid arthritis (Takatsu et al. 2005), hypertension (Brant et al. 1996; Agrawal et al. 2008), overweight and obesity (Fransen et al. 2008), larger waist circumference (Hwang et al. 2009), smoking (Agrawal et al. 2008), and lower levels of physical activity (Kiely et al. 2012). Environmental factors that are associated with increased prevalence include exposure to excessive noise (Van Eyken et al. 2007), ototoxic chemicals (Van Eyken et al. 2007), and ototoxic medications (Van Eyken et al. 2007). Fewer studies have prospectively examined risk factors for incident hearing impairment. Thus, it remains uncertain whether many of the factors found to be related to prevalence in cross-sectional studies are also prospectively associated with incidence and/or progression of hearing impairment.

Although there may be complex relations and/or interactions between many of these factors, for simplicity it is helpful to consider them according to four general categories: age; environment, such as exposure to occupational noise, recreational noise, or ototoxins; genetic predisposition, including sex and race; and medical and lifestyle factors, such as obesity, hypertension, diabetes, cardiovascular disease, diet, physical activity, smoking, or alcohol.

2.6.1 Age

Age is the strongest predictor of hearing impairment. In cross-sectional studies based on NHANES data, the prevalence of hearing impairment, defined as best ear $PTA_{(0.5,1,2,4\,kHz)}$ >25 dB HL, ranged from 0.6 % among 20–29 year olds to 63 % among those aged 70 and older (Agrawal et al. 2008; Lin et al. 2011b) (see Table 2.1). In prospective studies, age was a strong predictor of the 5-year incidence of hearing impairment (Cruickshanks et al. 2003; Mitchell et al. 2011). In a Dutch cohort, age was found to be the strongest nonaudiometric predictor of the rate of deterioration of hearing thresholds and the rate of deterioration was related to the square of baseline age (Linssen et al. 2014). Specifically, age explained 55 % (95 % CI 51–58 years) of the total interindividual variation in rate of deterioration. These findings are similar to those of other studies (Gates and Cooper 1991; Pearson et al. 1995; Viljanen et al. 2007; Kiely et al. 2012).

As age is such a strong risk factor for hearing impairment, the terms age-related hearing impairment (ARHI), age-related hearing loss (ARHL), and presbycusis have often been used interchangeably to describe the decline in hearing function observed with advancing age.

2.6.2 Environment

2.6.2.1 Noise Exposure

The relationship between excessive noise exposure and hearing impairment has been widely examined, and hearing impairment has been found to be associated with both occupational and leisure-time noise exposure (Rosenhall et al. 1990; Nondahl et al. 2000; Fransen et al. 2008; Agrawal et al. 2009). Although earlier studies have relied on a threshold shift at 4 kHz ("noise notch") as an indication of noise-induced hearing impairment (NIHL), more recent investigations suggest that this may not be a reliable measure (Nondahl et al. 2009). Overall, population-based studies of NIHL are scarce and interpretation may be hampered by a lack of validated measures for quantifying noise exposure over time and differences in measures used (Cruickshanks et al. 2010).

Hazardous noise exposure and NIHL can occur at any age. The NIDCD estimates that approximately 15 % of individuals in the United States between the ages of 20 and 69, corresponding to 26 million Americans, display hearing impairment that may have been caused by exposure to noise at work or during leisure-time activities (NIDCD 2014). A study that estimated the prevalence and evaluated the associated risk factors for noise-induced threshold shift (NITS) in the US adult population based on data from the NHANES found that the prevalence of unilateral, bilateral, and total NITS was 9.4, 3.4, and 12.8 %, respectively, and older age, male sex, and smoking were associated with higher odds of NITS (Mahboubi et al. 2013). Moreover, approximately 16 % of US adolescents aged 12–19 have reported some degree of hearing impairment, possibly related to hazardous noise exposure (Shargorodsky et al. 2010a; Clearinghouse 2014). Notably, noise exposure may increase vulnerability to ARHI (Gates et al. 2000;

Kujawa and Liberman 2006). Although anyone who is exposed to hazardous noise is at risk for hearing injury, the susceptibility to NIHL appears to vary. Factors such as smoking (Palmer et al. 2004; Wild et al. 2005), male sex, race, dietary factors, diabetes, cardiovascular disease (Daniel 2007), and exposure to carbon monoxide (Fechter 2004) may be associated with an increased risk of NIHL. In a Finnish study, a history of noise exposure was associated with worse high-frequency hearing thresholds only among those individuals with other concomitant otologic risk factors, such as ear infection, otosclerosis, Ménière's disease, sudden sensorineural hearing loss, ear or head trauma, or ototoxic medication use. No association between history of noise exposure and audiogram pattern was observed among those without other otologic risk factors, suggesting that certain otologic risk factors may "sensitize" the ear to noise-induced injury (Hannula et al. 2012).

The United States Centers for Disease Control and Prevention (CDC) reports that occupational hearing impairment is the most common work-related illness in the United States, with approximately 22 million US workers exposed to hazardous levels of workplace noise. Worker's compensation for hearing impairment disability accounts for $242 million in annual expenditures. The CDC also reports that in the year 2007, 14 % of reported occupational illness was due to NIHL, the vast majority of which was in the manufacturing sector (National Institute for Occupational Safety and Health 2011). Nonoccupational hazardous noise exposure, or leisure-time noise exposure, may be even more prevalent than occupational hazardous noise exposure (Clark 1991). Recreational firearm noise is a well-recognized cause of NIHL (Clark 1991).

In the pediatric population, noise exposures have been identified that are significantly associated with higher risk of hearing impairment. These include more than 4 hours per week of personal headphone use; more than 5 years of personal headphone use; more than 4 visits per month to a music club or discotheque; and residence on a mechanized farm (Vasconcellos et al. 2014).

2.6.2.2 Chemical Exposure

The CDC estimates that every year 10 million US workers are exposed to potentially ototoxic chemicals in the workplace (National Institute for Occupational Safety and Health 2015). Toluene, tricholorethylene, styrene, and xylene have all been implicated as potential ototoxins (Johnson and Nylen 1995; Fuente and McPherson 2006). Some studies observed an interaction between ototoxic chemical exposure and NIHL (Morata 2002; Fuente and McPherson 2006), suggesting that exposure to organic solvents in combination with excessive noise exposure multiplicatively increases the risk of hearing impairment (Fechter 2004).

2.6.2.3 Environmental Toxin Exposure

A cross-sectional study in adults aged 20–69 years based on NHANES data from 1999 to 2004 found a higher prevalence of audiometric hearing impairment among individuals with low-level exposure to cadmium or lead (Choi et al. 2012). In a study of 458 men who were participants in the VA Normative Aging Study, a cross-sectional association between higher bone lead levels (a marker of cumulative lead exposure) and poorer hearing thresholds was observed. In addition, higher bone lead levels were longitudinally associated with a greater deterioration in hearing thresholds, suggesting that chronic low-level lead exposure is a risk factor for hearing decline (Park et al. 2010).

2.6.2.4 Ototoxic Medications

The overall incidence of medication-related ototoxicity is not known. More than 130 different medications have been reported to be potentially ototoxic and administration of more than one medication with ototoxic potential may lead to multiplicative effects. Given that many potentially ototoxic medications are eliminated by the kidney, renal impairment is a risk factor for ototoxicity. Following is a discussion of some of the more common examples of potentially ototoxic medications.

Aminoglycoside antibiotics such as gentamicin, streptomycin, amikacin, neomycin, and kanamycin can be toxic to the cochlea and the stria vascularis. A higher risk of aminoglycoside-associated ototoxicity is found with extremes of age, a family history of aminoglycoside-related ototoxicity, aminoglycoside therapy that exceeds 2 weeks in duration, and antibiotic peak and trough concentrations that exceed therapeutic levels. Congenital hearing impairment has been associated with in utero exposure to kanamycin or streptomycin following maternal treatment with these antibiotics during pregnancy (Cunha 2001).

Loop diuretics may adversely affect the potassium gradient of the stria vascularis as well as the endocochlear electrical potential. Otoxicity associated with loop diuretics is usually dose related and thus more likely in the setting of reduced renal function owing to the accumulation of the medication. Ototoxicity due to furosemide is often reversible. Currently, ethacrynic acid is rarely used, in part because of its potential ototoxicity.

The chemotherapeutic agent cisplatin is ototoxic. In the treatment of pediatric cancer, higher risk for hearing impairment may be associated with the combination of cisplatin therapy and carboplatin therapy, radiotherapy, younger age at diagnosis, and genetic predisposition (Yasui et al. 2014). Several experimental approaches involving the concomitant administration of potentially otoprotective factors have been explored. However, their effectiveness has not been demonstrated (Rybak and Ramkumar 2007).

Ibuprofen, acetaminophen, and aspirin are the three most commonly used drugs in the United States. Potential ototoxicity due to high doses of salicylates and nonsteroidal anti-inflammatory drugs (NSAIDs) has been well described (Jung et al. 1993). In a large prospective cohort of men, the Health Professionals Follow-up Study (HPFS), regular use, two or more times per week, of the NSAIDs acetaminophen or aspirin was associated with an increased risk of hearing impairment (Curhan et al. 2010). The magnitudes of the associations with all three types of analgesic were greater in those younger than age 50: regular users of aspirin were 33 % more likely, regular users of NSAIDs were 61 % more likely, and regular users of acetaminophen were 99 % more likely to have hearing impairment than nonregular users of the same age. Regular moderate use of ibuprofen or acetaminophen 2 days per week or more was also associated with an increased risk of hearing impairment in a large prospective cohort of women, the Nurses' Health Study II (NHS II), but aspirin use was not (Curhan et al. 2012). The magnitude of the risk related to ibuprofen and acetaminophen use tended to increase with increasing frequency of use.

2.7 Genetic Predisposition

Heritability studies using twins and longitudinal studies of family cohorts have demonstrated that a substantial proportion of hearing impairment risk can be attributed to genetic predisposition, with heritability indices of 0.35–0.55 (Karlsson et al. 1997; Gates et al. 1999; Christensen et al. 2001). Differences in susceptibility due to gene–environment interactions may contribute to the considerable variation observed in age of onset, severity, pattern, and progression of hearing impairment. Differences attributable to genetic variation accounted for 75 % of the total variance in the better ear $PTA_{(0.5\text{–}4\ \mathrm{kHz})}$ among older women (Viljanen et al. 2007) and 66 % among older men (Wingfield et al. 2007). Some studies have identified genes potentially involved in common forms of hearing impairment. However, no confirmatory studies have yet been published. For example, a cross-sectional family study identified a possible locus for an age-related hearing impairment trait (Huyghe et al. 2008). Other studies have suggested potential susceptibility genes for noise-induced hearing impairment (Konings et al. 2009) and hearing impairment in older people (Uchida et al. 2011). Although heritability studies indicate an important role of genetics in the development of acquired hearing impairment, the specific genetic determinants remain to be elucidated.

2.7.1 Sex

Male sex has been associated with a higher overall prevalence of hearing impairment in a number of studies (Cruickshanks et al. 1998; Agrawal et al. 2008), particularly for high frequencies. In prospective studies, male sex was a strong predictor of higher 5-year incidence of hearing impairment (Cruickshanks et al. 2003; Mitchell et al. 2011). In a Dutch cohort, the rate of deterioration in hearing thresholds was 1.1 dB (95 % CI 0.8, 1.4) per decade faster for men than for women (Linssen et al. 2014).

2.7.2 Skin and Eye Pigmentation

The risk of hearing impairment is considerably lower for black than for white individuals, and the prevalence among those of Hispanic descent falls in between (Helzner et al. 2005; Agrawal et al. 2008). Cross-sectional data from the 2003–2004 NHANES, based on 1,258 adults aged 20–59 years who had assessment of Fitzpatrick skin type (skin color) and pure-tone audiometric testing, showed an association between race/ethnicity and hearing thresholds: black participants demonstrated the best hearing thresholds, followed by Hispanics, and then white individuals. However, these associations were not significant in analyses stratified by Fitzpatrick skin type. In analyses stratified by race, darker-skinned Hispanics had better hearing than lighter-skinned Hispanics by an average of 2.5 dB and 3.1 dB for speech and high-frequency PTA, respectively. However, there were no associations between level of pigmentation and hearing level among black or among white individuals (Lin et al. 2012). It has been hypothesized that skin pigmentation provides a marker of melanocyte function and differences in cellular melanin in the ear may contribute to observed differences in hearing impairment prevalence.

Information on the association between eye pigmentation and hearing levels is limited. Individuals with blue eye color may be more susceptible to noise-induced hearing damage (Carlin and McCroskey 1980), and younger males with black skin and brown eye color may be least susceptible (Kleinstein et al. 1984). Individuals with lighter eye color may be more susceptible to meningitis-related hearing impairment (Cullington 2001) and individuals with brown eyes may be more susceptible to cisplatin ototoxicity (Barr-Hamilton et al. 1991). However, the relationship between eye pigmentation and hearing remains unclear.

2.8 Medical and Lifestyle Factors

2.8.1 Infection

Up to two-thirds of children in the United States experience at least one episode of acute otitis media by the age of 3 years. Chronic otitis media (COM) is associated with both conductive and sensorineural hearing impairment. Population-based estimates of the prevalence and definitions of COM have varied. A study in the United Kingdom found that the prevalence of active and inactive COM was 4.1 % (Browning and Gatehouse 1992). Risk factors that may increase the risk for otitis media-related sensorineural hearing impairment include increasing age, longer duration, ear suppuration, size of tympanic membrane perforation, ossicular involvement, type of retraction, and radiographic evidence of soft tissue in the antrum and the round window niche (Yang et al. 2014).

2.8.2 Ménière's Disease

Although Ménière's disease can occur at any age, peak incidence occurs between the age of 40 and 60 years. According to NIDCD, approximately 615,000 individuals in the United States carry a diagnosis of Ménière's disease and 45,500 cases are newly diagnosed each year (Harris and Alexander 2010). However, incidence and prevalence estimates for Ménière's disease have varied considerably, possibly due to changes in the diagnostic criteria over time, as well as differences in the methodologies and populations studied. Reported prevalence in the United States ranges from 3.5 per 100,000 to 513 per 100,000 individuals. One study of the prevalence of Ménière's disease in a US population with health insurance found that the prevalence was 190 per 100,000 individuals, increased with increasing age, and demonstrated a female-to-male ratio of 1.9:1 (Alexander and Harris 2010).

2.8.3 Otosclerosis

The NIDCD estimates that more than three million US adults are affected by otosclerosis, and white, middle-aged women appear to be at higher risk. Prevalence estimates of otosclerosis vary depending on whether studies were based on clinically defined or histologically defined disease. An older US study found that the prevalence of otosclerosis was 0.5 % (Moscicki et al. 1985). In a Finnish study, the prevalence of otosclerosis among adults aged 62 and older was 1.3 % (Hannula et al. 2012) and in the United Kingdom, the prevalence among adults aged 41–60 years was 2.2 % (Browning and Gatehouse 1992). Both genetic and environmental factors may contribute to the development of the disease. Although several genes have been identified, the underlying pathophysiology remains unclear (Bittermann et al. 2014).

2.8.4 Cardiovascular Disease and CVD Risk Factors

Reduced blood supply to the cochlea, whether due to microvascular or macrovascular compromise, can lead to capillary constriction within the stria vascularis, cell death, and poorer hearing sensitivity (Gates et al. 1993; Torre et al. 2005; Liew et al. 2007). Therefore, factors that may compromise cochlear blood supply or the cochlear microcirculation may adversely influence hearing function. A few studies have examined cardiovascular disease (CVD) as a risk factor for hearing impairment (Rubinstein et al. 1977; Gates et al. 1993; Torre et al. 2005), while potential associations between specific CVD risk factors, such as plasma lipids, blood pressure and smoking, and hearing thresholds have been examined in several cross-sectional studies (Gates et al. 1993; Evans et al. 2006; Helzner et al. 2011) and some prospective studies (Shargorodsky et al. 2010c; Gopinath et al. 2011a; Simpson et al. 2013). Findings from studies of the association between CVD as well as CVD risk factors and the risk of hearing impairment have been inconsistent.

Cross-sectional associations between poorer hearing sensitivity and CVD events have been observed in women and less consistently in men. In one study of older individuals, the odds ratio for hearing loss among women with any CVD event was 3.06 (95 % CI: 1.84–5.10); the odds ratio for hearing loss among men with coronary heart disease was 1.68 (95 % CI, 1.10–2.57) and among men with history of stroke was 3.46 (95 % CI: 1.60–7.45) (Gates et al. 1993). In the EHLS, women with self-reported history of myocardial infarction (MI) were twice as likely to have hearing impairment as those with no history of MI (95 % CI: 1.15–3.46). However, no association was observed for men (Torre et al. 2005).

Several large cross-sectional studies have observed associations between specific CVD risk factors and hearing impairment. However, results have been inconsistent. For example, in data from NHANES 1999–2004, hypertension and diabetes were associated with a higher prevalence of high-frequency hearing impairment (Agrawal et al. 2008; Bainbridge et al. 2008). A study in Sweden of older individuals observed cross-sectional associations between high systolic blood pressure and low- to mid-frequency hearing impairment in women older than 79 years and between high diastolic blood pressure and low- to midfrequency hearing impairment in women aged 85 years. However, no associations were observed in women younger than age 79 or in men aged 70–85 (Rosenhall and Sundh 2006). In a prospective study of 531 men in the Baltimore Longitudinal Study of Aging (BLSA), a 32 % higher risk of hearing impairment was observed for each 20 mmHg higher systolic blood pressure (Brant et al. 1996). In a large prospective study of cardiovascular risk factors among an older cohort of men (HPFS), hypertension was not significantly associated with risk of self-reported hearing impairment (Shargorodsky et al. 2010c).

Although a higher prevalence of hearing impairment has been observed among individuals with diabetes in some studies, findings have been inconsistent. In one study, sensorineural hearing impairment was observed to be more common among type 1 diabetic patients than among age-matched controls (Kakarlapudi et al. 2003). In a different study, high-frequency hearing impairment was observed to be more common among diabetic individuals up to age 60 but not among older individuals (Vaughan et al. 2006). A NHANES study found a twofold higher prevalence of hearing impairment among individuals with self-reported diabetes (Bainbridge et al. 2008). A meta-analysis of 13 cross-sectional studies conducted between 1950 and 2011 found the overall pooled odds ratio of the prevalence of hearing impairment among diabetic individuals compared to nondiabetics was 2.2 (Horikawa et al. 2013). Nevertheless, a prospective association between diabetes and hearing impairment has not been observed in longitudinal studies (Shargorodsky et al. 2010c; Kiely et al. 2012).

Although several studies have examined the relationship between plasma lipids and hearing thresholds, no consistent associations have emerged (Lee et al. 1998; Gopinath et al. 2011a; Simpson et al. 2013). One case-control study with more than 4,000 cases of hearing impairment found a significant association between hyperlipidemia and NIHL (Chang et al. 2007). However, no significant associations between lipid levels and hearing sensitivity were observed in a prospective study of 837 older adults (Simpson et al. 2013) nor were significant associations observed between plasma cholesterol, high-density lipoprotein (HDL), or triglycerides in several other cross-sectional or case-control studies (Jones and Davis 1999, 2000; Gopinath et al. 2011a).

A number of other CVD risk factors have been examined. In the Health ABC Study, high triglyceride levels, higher resting heart rate, and smoking were associated with poorer hearing sensitivity in men; higher body mass index, higher resting heart rate, faster aortic pulse-wave velocity, and lower ankle–arm index were associated with poorer hearing sensitivity in women (Helzner et al. 2011). In the Beaver Dam Offspring Study (BOSS), larger central retinal venular equivalent (CRVE) and carotid intima media thickness, indicators of systemic cardiovascular disease, were associated with poorer hearing thresholds (Nash et al. 2011).

Acute hearing impairment has been associated with approximately 10 % of posterior circulation ischemic strokes (Lee 2014). Although these individuals often have at least partial or complete hearing recovery, a longitudinal study found that having two or more risk factors for stroke, such as hypertension, diabetes, smoking, or hyperlipidemia, and profound hearing impairment were inversely associated with hearing recovery (Kim et al. 2014).

Isolated acute hearing impairment was the initial presenting symptom in 31 % of cases of vertebral artery ischemic stroke, occurring up to 10 days before onset of additional symptoms (Lee et al. 2005). A cross-sectional analysis in the Blue Mountains Hearing Study (BMHS) found that individuals with moderate-to-severe hearing impairment were twice as likely to report a history of previous stroke as individuals without hearing impairment. However, no prospective association between moderate-to-severe hearing impairment and incidence of stroke after 5 years of follow-up was observed, although the study power was limited (Gopinath et al. 2009).

Several small studies have indicated a higher prevalence of high-frequency hearing impairment in individuals with chronic kidney disease (CKD). For example, in one study of older individuals with CKD, the prevalence of high-frequency hearing impairment was higher in those with CKD (39 %) than in age-matched controls (23 %) (Antonelli et al. 1990).

Few studies have investigated the relationship between systemic inflammation and hearing impairment. A prospective study in the EHLS found that among individuals younger than 60 years, those with consistently high C-reactive protein (CRP) levels (>3 mg/L) or whose CRP levels increased with time were almost twice as likely to develop hearing impairment over the 10-year follow-up as those without elevated CRP levels. No association was observed among individuals older than 60 years. Baseline measures of CRP, interleukin-6, or tumor necrosis factor-alpha were not associated with risk (Nash et al. 2014). In contrast, another prospective study among older adults found no association between CRP levels and change in hearing thresholds over time (Simpson et al. 2013).

Some autoimmune and inflammatory conditions have been associated with higher risk of hearing impairment. Autoimmune inner ear disease (AIED) or immune-mediated sensorineural hearing impairment accounts for less than 1 % of all hearing impairment in the United States. Of those with AIED, approximately 20 % have another autoimmune disease such as rheumatoid arthritis or systemic lupus erythematosus.

Vasculitis and some hematologic disorders may result in acute hearing impairment because of alterations in the microcirculation of the inner ear and hypercoagulability. For example, sickle cell anemia, polycythemia, leukemia, macroglobulinemia, and Berger's disease have all been associated with acute hearing impairment. Acute hearing impairment due to vascular compromise is often associated with vestibular dysfunction and vertigo that may not be related to the severity of hearing impairment.

In cross-sectional studies, higher body mass index (BMI), a measure of overall obesity, and larger waist circumference, a measure of central adiposity (Ketel et al. 2007), have been associated with poorer hearing thresholds (Fransen et al. 2008; Hwang et al. 2009; Helzner et al. 2011). In a large prospective study of US women (NHS II), higher BMI and larger waist circumference were independently associated with an increased risk of self-reported hearing impairment (Curhan et al. 2013). Larger waist circumference was independently associated with increased risk of hearing impairment even after adjusting for BMI, similar to previous cross-sectional findings (Hwang et al. 2009), suggesting that central adiposity may itself be a risk factor for hearing impairment.

Small cross-sectional studies have reported associations between higher levels of physical activity, higher cardiorespiratory fitness, and better hearing sensitivity (Hutchinson et al. 2010; Loprinzi et al. 2012). In a large prospective study of US women (NHS II), a higher level of physical activity was independently associated with a 17 % lower risk of hearing impairment. The risk decreased with increasing level of physical activity; even walking regularly was associated with a 15 % lower risk of hearing impairment (Curhan et al. 2013).

2.8.5 Obstructive Sleep Apnea

Few studies have examined the potential relationship between obstructive sleep apnea (OSA) and hearing impairment. A small cross-sectional study of individuals who snore found significantly elevated pure-tone hearing thresholds, lower distortion product otoacoustic emissions (DPOAE) amplitudes, and smaller brainstem auditory evoked potentials for those with OSA than for those without OSA, suggesting that OSA may be a risk factor for auditory dysfunction (Casale et al. 2012).

2.8.6 Preeclampsia

A small cross-sectional study of 40 pregnant women with preeclampsia and 30 pregnant women without preeclampsia found a higher likelihood of otoacoustic emissions abnormalities among women with preeclampsia. This suggests that preeclampsia may be associated with a higher risk of cochlear damage and hearing impairment (Baylan et al. 2010).

2.8.7 Hormonal Factors

Estrogen receptors are present in the inner ear, and human studies have suggested associations between low serum estrogen levels, such as in menopause, and hearing impairment (Jonsson et al. 1998). Fluctuations in hearing sensitivity have been demonstrated throughout the menstrual cycle (Swanson and Dengerink 1988), with reduced hearing during the menstrual phase when estrogen levels are lowest, a phenomenon not observed in women taking oral contraceptives. A cross-sectional study of the association between serum estradiol level and hearing sensitivity in 1,830 postmenopausal women demonstrated a lower prevalence of hearing impairment among those with higher levels of estradiol (Kim et al. 2002).

In postmenopausal women, the prevalence of ARHI was observed to be lower in those who were taking estrogen (Hultcrantz et al. 2006; Hederstierna et al. 2007). In contrast, progesterone or progestin may adversely influence hearing in women, whether during the luteal phase of the normal menstrual cycle or as part of postmenopausal hormone (PMH) use (Guimaraes et al. 2006). In a cross-sectional study of 109 postmenopausal women, 20 of whom used estrogen therapy alone, 30 combined estrogen plus progestin, and 59 used no PMH, women who used estrogen therapy alone had significantly lower mean air conduction thresholds than those who used combined PMH or no PMH (Kilicdag et al. 2004). A cross-sectional study of 143 women in Sweden found that postmenopausal women who did not use PMH had poorer hearing sensitivity than women who were pre- and perimenopausal or those who were postmenopausal but did use PMH therapy (Hederstierna et al. 2007). A study of 124 postmenopausal women who had used either estrogen alone ($n=30$), estrogen plus progestin ($n=32$), or no PMH ($n=62$), found that women who used combined estrogen plus progestin had poorer hearing and poorer speech perception in background noise than those who used estrogen alone or no PMH. However, no differences were seen in hearing sensitivity between those who used estrogen alone and those who used no PMH (Guimaraes et al. 2006).

Aldosterone has a stimulatory effect on expression of sodium–potassium ATPase and the sodium–potassium–chloride cotransporter in cell membranes. A cross-sectional study observed that a higher serum aldosterone level, yet within the normal clinical range, was associated with better pure-tone hearing thresholds and better performance on the hearing in noise test, suggesting that aldosterone may have a protective influence on peripheral auditory function (Tadros et al. 2005). However, no associations were observed with transient evoked otoacoustic emissions (TEOAE) or gap detection.

2.8.8 Dietary Factors

The associations between dietary intake patterns or individual nutrient intakes and hearing have been examined in several cross-sectional studies. However, prospective information on the association between dietary factors and the risk of developing hearing impairment is more limited. A cross-sectional study in NHANES found that higher overall dietary quality was associated with better pure-tone hearing thresholds at high frequencies but not at low frequencies (Spankovich and Le Prell 2013).

It has been proposed that higher intake of antioxidant nutrients, such as vitamin A and carotenoids, vitamins C and E, and folate, may protect against oxidative stress and cochlear damage (Seidman 2000; Seidman et al. 2004; Darrat et al. 2007).

A recent metabolomics and network analysis identified the retinoic acid pathway as a promising target for the development of prevention and treatment strategies (Muurling and Stankovic 2014). In animal models, vitamin C, vitamin E, and beta-carotene have been shown to be protective against hearing impairment (Seidman 2000; Takumida and Anniko 2005; Le Prell et al. 2007). In humans, findings from cross-sectional studies of the relationship between intake of vitamins A, C, and E and carotenoids and risk of hearing impairment have been inconsistent and prospective data are limited (Gopinath et al. 2011b; Spankovich et al. 2011; Peneau et al. 2013). Although some significant cross-sectional associations were observed in the BMHS, no longitudinal association was observed between dietary intake of vitamins A, C, and E or beta-carotene and 5-year incidence of hearing impairment (Gopinath et al. 2011b). Similarly, a prospective study of over 26,000 older men in the HPFS did not observe an association between intakes of vitamins C or E or beta-carotene and risk of self-reported hearing impairment (Shargorodsky et al. 2010b). In a European interventional trial, women with higher intake of vitamin B_{12} had better hearing thresholds than those with lower intake, but no associations were observed with retinol; beta-carotene; folate; or vitamins B_6, C, or E; no associations were observed for men (Peneau et al. 2013). Cross-sectional studies suggest a relation between both low intake and low plasma red blood cell levels of folate and higher prevalence of hearing impairment. However, the studies were small and the results inconsistent (Houston et al. 1999; Berner et al. 2000). A randomized clinical trial in the Netherlands, a country without folate fortification of the food supply, found that daily oral folic acid supplementation slowed hearing decline in the speech-frequency range (Durga et al. 2007).

Fish consumption and higher intake of long-chain omega-3 polyunsaturated fatty acids (LC omega-3 PUFA) may help maintain adequate cochlear blood flow and protect against ischemic injury. In addition, the LC omega-3 PUFA, specifically eicosapentaenoic acid (EPA) (20:5ω-3) and docosahexaenoic acid (DHA) (22:6ω-3), as found in fish, may have a beneficial influence on membrane structure and function, gene expression, and proinflammatory and prothrombotic factors (Mozaffarian and Wu 2011). In the BMHS, the 5-year incidence of hearing impairment was 42 % lower among individuals who consumed two or more servings of fish per week than among those who consumed less than one serving of fish per week. Higher intake of LC omega-3 PUFA was also inversely associated with the 5-year incidence of audiometrically measured hearing impairment (relative risk = 0.76) (Gopinath et al. 2010a). A cross-sectional European study found that higher seafood and shellfish intake was associated with better hearing thresholds in men but not in women (Peneau et al. 2013). In a prospective study of more than 65,000 women in the NHS II, consumption of two or more servings of fish per week was associated with lower risk of self-reported hearing loss. In comparison with women who rarely consumed fish (less than one serving per month), the multivariable-adjusted relative risk (RR) for hearing loss among women who consumed two to four servings of fish per week was 0.80 (p-trend < 0.001), and higher consumption of each specific fish type was inversely associated with risk (p-trend ≤ 0.04). Higher intake of long-chain omega-3 PUFA was also inversely associated with risk; in comparison with women in the lowest quintile of intake of long-chain omega-3 PUFA, the multivariable-adjusted RR for hearing loss among women in the highest quintile was 0.85 and in the highest decile was 0.78 (p-trend $<$ 0.001) (Curhan et al. 2014).

2.8.9 Alcohol Intake

Moderate alcohol intake may protect cochlear blood flow (Seidman et al. 1999) and directly enhance neuroprotective mechanisms that preserve hearing (Collins et al. 2009). However, chronic excess alcohol intake has been associated with irreversible hearing impairment (Rosenhall et al. 1993). Acute alcohol intake may temporarily impair auditory processing and worsen auditory thresholds (Robinette and Brey 1978; Fitzpatrick and Eviatar 1980; Hienz et al. 1989; Pearson et al. 1999; Liu et al. 2004; Kahkonen et al. 2005; Upile et al. 2007) and may also adversely alter central processing of auditory information (Fitzpatrick and Eviatar 1980; Meerton et al. 2005; Upile et al. 2007). Some evidence suggests that long-term moderate alcohol intake may protect against hearing impairment (Popelka et al. 2000; Gopinath et al. 2010b; Dawes et al. 2014). In humans, some cross-sectional studies reported an inverse association between moderate alcohol consumption and hearing impairment (Popelka et al. 2000; Gopinath et al. 2010c), although others did not (Brant et al. 1996; Sousa et al. 2009). In a prospective study of 870 men and women age 49 and older (BMHS), no association was observed between alcohol consumption and the 5-year incidence of measured hearing impairment. However, there was insufficient power to be conclusive (Gopinath et al. 2010c). In a prospective study of more than 26,000 older men (HPFS), no association between moderate alcohol consumption and the risk of self-reported hearing impairment was observed (Curhan et al. 2011).

2.8.10 Smoking and Tobacco Use

Smoking has been associated with a higher risk of hearing impairment in several studies (Rosenhall et al. 1993; Cruickshanks et al. 1998; Uchida et al. 2005; Dawes et al. 2014) but not all (Brant et al. 1996; Itoh et al. 2001; Helzner et al. 2005; Fransen et al. 2008). In the EHLS, current smokers were 1.7 times more likely to have hearing impairment. Notably, there is strong evidence that even low levels of passive tobacco exposure, both in utero and due to secondhand smoke exposure during childhood, are associated with higher risk of sensorineural hearing impairment in the pediatric population (Korres et al. 2007; Durante et al. 2011; Lalwani et al. 2011). Based on NHANES data, prenatal smoke exposure was significantly associated with elevated pure-tone hearing thresholds at 2 and 6 kHz and a 2.6 times higher odds of unilateral low-frequency hearing impairment in adolescents whose mothers reported smoking during pregnancy (Weitzman et al. 2013).

2.8.11 Level of Education and Type of Occupation

In the EHLS, individuals with fewer years of education were more likely to develop hearing impairment than those with 16 or more years of education (Cruickshanks et al. 2003). Participants who worked in industrial jobs were almost twice as likely to develop hearing impairment as participants who had management or professional employment. In BOSS, individuals with 12 or fewer years of education were almost twice as likely to have hearing impairment as individuals with 16 or more years of education. Those who reported occupational noise exposure were also at higher risk (Nash et al. 2011). This is consistent with cross-sectional findings based on NHANES data that showed a higher prevalence of hearing impairment among individuals with fewer years of education (Agrawal et al. 2008). A study in Brazil found that individuals with an occupation related to agriculture, industry, or maintenance were at higher risk for self-reported hearing impairment (Cruz et al. 2013).

2.8.12 Hearing Impairment and Dementia

A prospective study of older adults in Utah observed that, after adjusting for sex, presence of the *APOE-e4* allele, education, and baseline age, hearing impairment was independently associated with 28 % higher risk of dementia (Gurgel et al. 2014). Findings for dementia were similar in the BLSA and also showed that hearing impairment was independently associated with lower scores on tests of memory and executive function (Lin et al. 2011c, d).

2.9 Sudden Sensorineural Hearing Loss

Sudden sensorineural hearing impairment (SSNHL) is typically defined as a >30 dB worsening in hearing thresholds across three contiguous frequencies that occurs within a 1- to 3-day period. It is estimated that between 5 and 20 per 100,000 individuals per year suffer from SSNHL. Factors reported to be associated with higher risk of SSNHL include hypertension, diabetes, heavy smoking, and heavy alcohol consumption (Lin et al. 2012). A study in Taiwan found that individuals who suffered from migraine headaches were almost twice as likely to develop SSNHL as those who did not (Chu et al. 2013). In addition, individuals with SSNHL may have lower levels of plasma folate (Cadoni et al. 2004) and nervonic acid, an omega-9 polyunsaturated fatty acid (Cadoni et al. 2010).

Genetic risk factors for SSNHL have been explored and findings suggest that the presence of certain thrombophilic factor polymorphisms may increase the risk of SSNHL. Genetic polymorphisms associated with factor V Leiden, prothrombin, and the MTHFR 677 enzyme have been associated with higher risk of SSNHL (Ballesteros et al. 2012).

2.10 Epidemiology of Pediatric Hearing Impairment

Hearing impairment is the most common birth defect in industrialized countries and the most prevalent sensorineural disorder. Between 1 and 3 out of every 1,000 infants born in the United States are affected by congenital hearing impairment (Kemper and Downs 2000), and 1 of every 500 newborns in industrialized countries has bilateral permanent sensorineural hearing impairment >40 dB HL (Hilgert et al. 2009). From birth to age 5, the prevalence increases to 2.7 per 1,000 and further increases to 3.5 per 1,000 in adolescents (Morton and Nance 2006).

The NIDCD reports that 9 out of 10 infants who are born deaf are born to hearing parents. In developed countries, it is estimated at least two-thirds of prelingual cases of hearing impairment are attributable to identified genetic causes, and the remaining one-third of cases are attributable either to environmental factors or to not yet identified genetic factors. The most common environmental cause of congenital hearing impairment is congenital infection with cytomegalovirus (CMV), an infection with overall birth prevalence of 0.64 %, the majority of which (>90 %) are asymptomatic infections. In developed nations, unilateral or bilateral hearing impairment develops in up to 4.4 % of children born with asymptomatic CMV by the age of 6 years. However, this varies with ethnicity and the hearing impairment may fluctuate (Kenneson and Cannon 2007). Other congenital infections that contribute to newborn hearing impairment include the TORCH infections, which refers to toxoplasmosis, “other” [e.g., syphilis, varicella-zoster virus or “chicken pox,” fifth disease or Parvovirus B19, and Human Immunodeficiency Virus (HIV)], rubella, CMV, and herpes simplex virus. In addition, bacterial infections, such as *Neisseria meningitidis*, *Haemophilus influenzae*, and *Streptococcus pneumoniae*, and meningitis due to infection with organisms such as *Escherichia coli*, *Listeria monocytogenes*, *Enterobacter cloacae*, or *Streptococcus agalactiae*, can lead to hearing impairment. Perinatal anoxia, hyperbilirubinemia, and ototoxic medications may also contribute to newborn hearing impairment.

A 2004 overview of the epidemiology of hearing impairment in US newborns, children, and adolescents found that the incidence of permanent childhood hearing impairment among newborns, as reported by 47 states, was 1.1 per 1,000 screened (3,600 with hearing impairment out of 3,496,452 screened; Mehra et al. 2009). Hearing impairment was defined as $PTA_{(0.5–2\ kHz)}$ >20 dB, unilateral or bilateral, and either sensorineural or nontransient conductive. The lowest incidence rate was found in North Dakota (0.22/1,000) and the highest in Hawaii (3.61/1,000).

In 2011, data from the CDC showed that newborn screening for hearing impairment was performed on almost 98 % of US-born infants, up from 46.5 % in 1999 (National Center on Birth Defects and Developmental Disabilities [NCBDDD] 2012). Of those infants who underwent screening, 1.8 % did not pass their most recent follow-up or final screening (NCBDDD 2009). Notably, birth weight was less than 2,500 g (~5.5 lbs.) in approximately one-fourth of infants with hearing impairment and approximately 25 % had another disability (e.g., vision impairment or cerebral palsy).

Data from studies conducted between 1958 and 1995 that used audiometric screening of children and adolescents showed that the average prevalence of unilateral or bilateral hearing impairment ($PTA_{(0.5,1,2\ kHz)}$ >25 dB) was 3.1 % (range 1.7–5.0 %) during this period. The average prevalence of mild or worse bilateral hearing impairment ($PTA_{(0.5,1,2\ kHz)}$ >25 dB) was 0.9 % (range, 0.4–1.7 %). The average prevalence of moderate or worse bilateral hearing impairment ($PTA_{(0.5,1,2\ kHz)}$ >40 dB) was 0.3 % (range, 0.11–0.74 %). Notably, the prevalence estimates for any hearing impairment as measured by surveys (1.9 %, ranging from 1.3 to 4.9 %) were similar to estimates for mild unilateral or bilateral hearing impairment provided by audiometric measurement (3.1 %, ranging from 1.7 to 5.0 %), suggesting that questionnaire or interview queries regarding hearing may be a simple and useful screening tool (Mehra et al. 2009). A study of adolescents aged 12–19 years based on NHANES that compared data from the 1988–1994 and 2004–2005 time periods demonstrated that the prevalence of any hearing impairment (unilateral or bilateral low- or high-frequency PTA >15 dB) increased significantly from 14.9 % in 1988–1994 to 19.5 % in 2005–2006. In 2005–2006, hearing impairment was more commonly unilateral and involved higher frequencies ($PTA_{(3,4,6,8\ kHz)}$; see Table 2.2). The prevalence of hearing impairment did not significantly differ by age or race/ethnicity in either time period. However, females were significantly less likely than males to demonstrate any hearing impairment in 2005–2006. In addition, adolescents from families living below the federal poverty threshold were 60 % more likely to have hearing impairment than those living above the poverty threshold (Shargorodsky et al. 2010a).

Overall, the most commonly identified risk factors for hearing impairment in children include genetic disorders and syndromes, prenatal infections, a family history of childhood hearing impairment, history of a neonatal intensive care unit stay longer than 5 days, craniofacial abnormalities, central nervous system diseases, exposure to ototoxic medications, and head trauma.

Based on an analysis of US studies with varying definitions of hearing impairment (Mehra et al. 2009), the etiology of bilateral, moderate, or worse sensorineural hearing impairment in US children and youth was unknown in 56 % of cases. Hearing impairment was determined to be genetic in 23 % (±13 %) of cases, of which 48 % of cases were nonsyndromic. Hearing impairment was determined to be acquired in 20 % (±7 %) of cases, of which 17 % of cases were prenatal, 12 % were perinatal, and 71 % were postnatal. In 1 % of cases, hearing impairment was attributable to other etiologies, such as posterior fossa tumors, cysts, and complications resulting from their removal; cochlear dysplasia; and congenital malformations of the ear. In a study of risk factors associated with risk of neonatal hearing impairment in Poland, the largest proportion, approximately 16 %, of sensorineural hearing impairment was found in infants with identified or suspected syndromes associated with hearing impairment. The largest proportion of neonatal sensorineural impairment occurred in the absence of any identified risk factors.

2.10.1 Demographic Factors

Hearing impairment is 10–40 % more common in male than female children in both audiometric-based and survey-based studies (Mehra et al. 2009). A NHANES study demonstrated that in adolescents aged 12–17 years, females were 24 % less likely than males to have any hearing impairment and 39 % less likely to have high-frequency hearing impairment (Shargorodsky et al. 2010a). Mexican-American children had a higher prevalence of hearing impairment than non-Hispanic white children. Based on NHANES data, the prevalence of hearing loss was higher among children living in households within the lowest stratum of family income or in poverty (Shargorodsky et al. 2010a).

2.10.2 Genetics and Pediatric Hearing Impairment

Of infants with hearing impairment detected on newborn hearing screenings, 30 % display other physical findings associated with syndromic hearing impairment; the remaining 70 % are considered to be nonsyndromic (Van Camp et al. 1997). Hearing impairment is an associated feature of more than 400 genetic syndromes, the most common of which include Usher (Toriello et al. 2004) (4 infants in every 100,000 births), Pendred, Jervell, and Lange-Nielsen syndromes. Most cases (~80 %) of inherited hearing impairment are monogenic and typically present prelingually. Approximately half of autosomal recessive nonsyndromic hearing impairment is attributable to the DFNB1 disorder caused by mutations in the *GJB2* gene that encodes connexin 26 and the *GJB6* gene that encodes connexin 30. The carrier rate in the general population for the *GJB2* gene mutations associated with inherited hearing impairment is 1 in 33 (Smith et al. 2014). In contrast, autosomal dominant inherited hearing impairment occurs in approximately 20 % of cases and often presents postlingually. Rarely, X-linked or mitochondrial inheritance can occur (Cryns and Van Camp 2004). Notably, monogenic hearing impairment is heterogeneous and more than 100 mapped loci and 46 potentially causal genes have been identified (Hilgert et al. 2009).

2.10.3 Congenital Hypothyroidism

Left untreated, congenital hypothyroidism can lead to hearing impairment that is typically bilateral, mild-to-moderate, sensorineural high-frequency hearing impairment. A study of young adults who were diagnosed and treated for congenital hypothyroidism found that a higher risk of hearing impairment was associated with the type of congenital hypothyroidism; the risk was twofold higher for hearing impairment among individuals diagnosed with athyreosis and gland in situ than for those with an ectopic gland (Lichtenberger-Geslin et al. 2013).

According to 2012 WHO estimates, 360 million individuals worldwide have hearing impairment, representing 5.3 % of the world's total population. Of these, 328 million adults have disabling hearing loss, defined as better ear $PTA_{(0.5,1,2,4\,kHz)}$ >40 dB HL, including 183 million males, 145 million females, and one-third of individuals older than the age of 65 years. In addition, 32 million children have disabling hearing loss, defined as better ear $PTA_{(0.5,1,2,4\,kHz)}$ >30 dB HL (WHO 2012). These estimates are based on analyses that synthesized data from 42 studies in 29 countries to determine the global and regional prevalence of hearing impairment as part of the Global Burden of Disease (GBD) project, an endeavor that provides cause-specific estimates of global mortality, disease burden and risk factors for fatal and nonfatal conditions (Stevens et al. 2013). Of these 42 studies, 18 were in high-income countries, 24 in low- or middle-income countries, 13 examined only children aged <20 years, and 17 examined individuals of all ages. Despite limited global data, the GBD findings illustrate that the prevalence of hearing impairment is considerable and adult onset hearing impairment is the third leading cause of disability.

The GBD study found that hearing impairment prevalence increased with age and was higher among males than females (see Figs. 2.1 and 2.2). The global prevalence of moderate or worse hearing impairment, when defined as better ear $PTA_{(0.5,1,2,4\,kHz)}$ >35 dB HL, was 12 % for males and 10 % for females aged 15 years and older. The prevalence of mild hearing impairment, defined as better ear $PTA_{(0.5,1,2,4\,kHz)}$ between 20 and 34 dB HL, was 23 % for adult males and 19 % for adult females. Among children aged 5–14 years, the global prevalence of moderate or worse hearing impairment was 1.4 %.

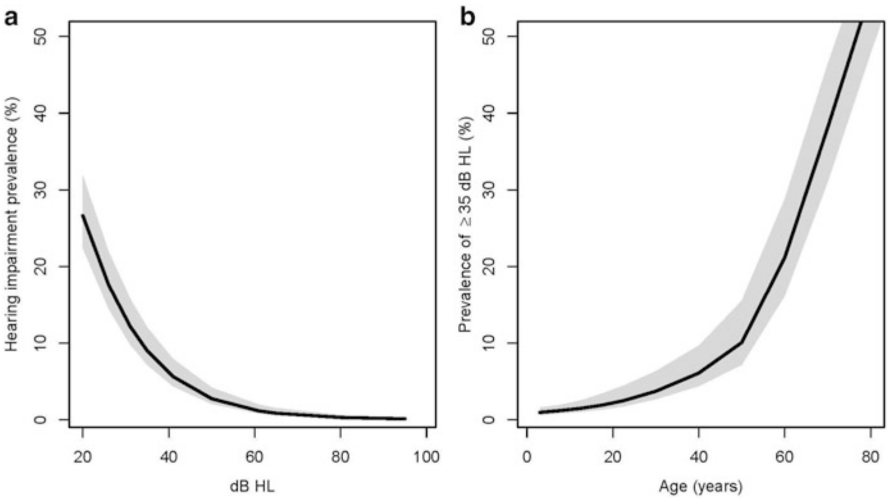

Fig. 2.1 Global pattern of hearing impairment (**a**) by hearing threshold and (**b**) by age. Age-standardized cumulative prevalence, that is, prevalence of hearing impairment at each threshold and at higher thresholds, is shown in (**a**). Solid lines show central estimates and shaded areas show 95 % uncertainty intervals [Copyright permission received; source *European Journal of Public Health,* 23(1), 146–152, 2013]

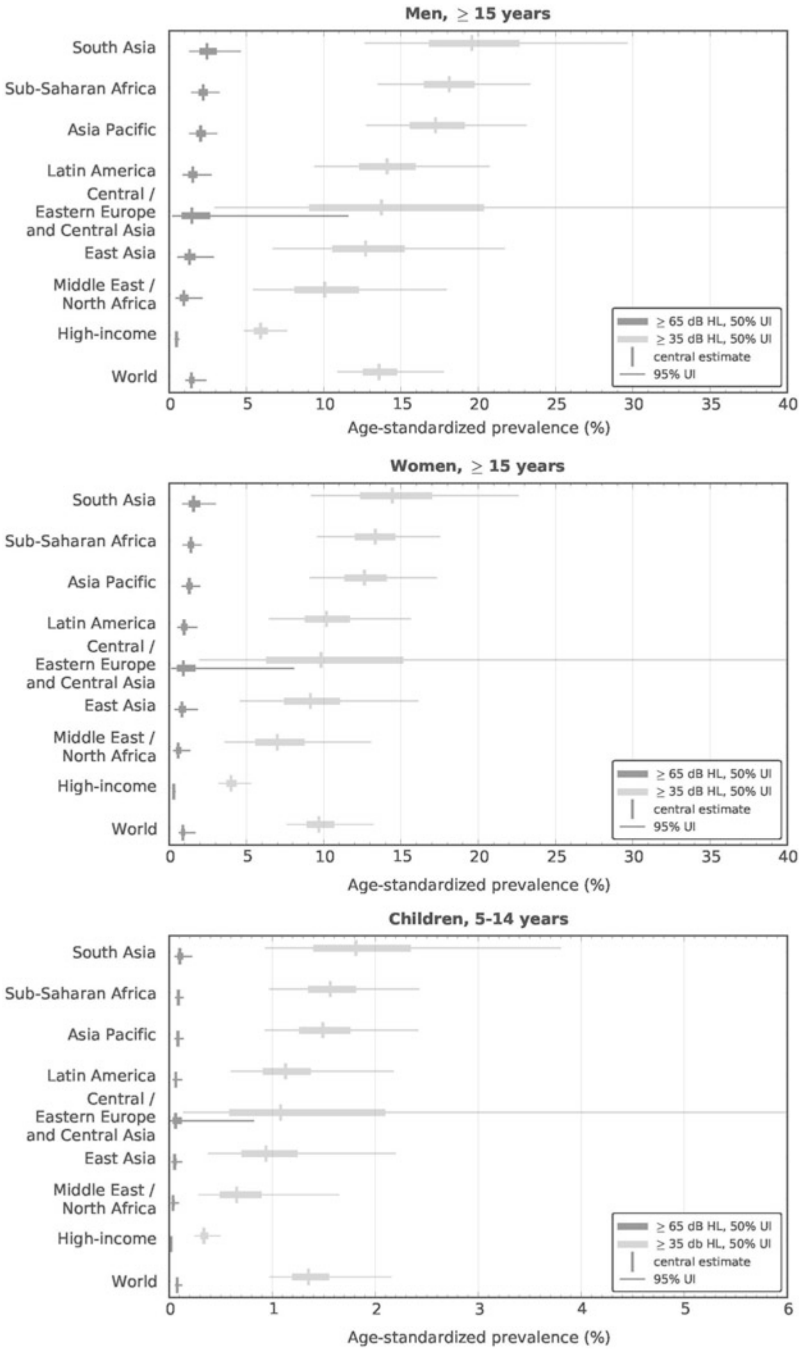

Fig. 2.2 Age-standardized prevalence of hearing impairment, 2008. Age-standardized prevalence of hearing impairment by region, men ≥15 years, women ≥15 years, and children 5–14 years of age. UI uncertainty interval [From *European Journal of Public Health,* 23(1), 146–152, 2013. Reprinted with permission.]

After adjusting for differences in age structure, the prevalence of adult hearing impairment was greatest in developing regions and lowest in high-income regions. For example, estimates show that the age-standardized prevalence of moderate or worse hearing impairment was fourfold higher in South Asia (13 %), the area with the greatest percentage of adults with hearing impairment, than in high-income regions. Areas with the lowest prevalence of moderate or worse hearing impairment among adults included the Middle East and North Africa region (6 %) and the high-income regions (8 %). The prevalence of moderate or worse hearing impairment in Central and Eastern Europe and Central Asia was 14 %. Among the 32 million children with disabling hearing impairment, the prevalence was greatest in South Asia, Asia Pacific, and Sub-Saharan Africa. In both children and adults, the prevalence of hearing impairment decreased exponentially as gross national income (GNI) increased. For most regions, the prevalence in children decreased linearly as parents' literacy rate increased (WHO 2012).

According to estimates from the WHO, the highest proportion of the total burden of overall global hearing impairment is attributable to genetic causes, otitis media, and ARHI (Mathers 2003). A moderate proportion is attributable to excessive noise exposure, ototoxic medication use, ototoxic chemical exposure, prenatal and perinatal complications, infectious disease, cerumen, or foreign bodies. A smaller proportion is attributable to nutritional deficiencies, trauma, Ménière's disease, tumors, or cerebrovascular disease. The major contributors to conductive hearing impairment include COM, both chronic suppurative otitis media and otitis media with effusion; tympanic membrane perforation; cholesteatoma; and otosclerosis. Adult onset hearing impairment is a leading cause of total global years lived with disability (YLDs); in 2001 YLDs were estimated to be 26 million, or 5 % of YLDs attributable to all causes (Mathers 2003; Mazelova et al. 2003).

The WHO world population estimates indicate that in the year 2025, 1.2 billion individuals will be older than 60 years, projecting to more than 500 million individuals worldwide expected to suffer from disabling hearing impairment (Sprinzl and Riechelmann 2010; WHO 2012). The consequences of disabling hearing impairment are considerable. The adverse impact on interpersonal communication, psychosocial well-being, economic independence, and overall quality of life can be substantial. In children, the detrimental effects on speech and language development can limit educational achievement and future employment opportunities and harm social-emotional development. Recognizing that the assessment of disability-related functional impairment must take the environmental context into account, in 2013 the Global Burden of Disease Hearing Loss Expert Group proposed a revised classification of hearing impairment that describes disabling hearing impairment as better ear $\mathrm{PTA}_{(0.5,1,2,\text{ and }4\text{ kHz})}$ $\geq$35 dB HL, either unilateral or bilateral, for all age groups. Based on this definition, approximately 538 million individuals worldwide aged older than 5 years have disabling hearing impairment. The apparent upward trend in the global prevalence of hearing loss may reflect an increased mean life expectancy in many countries and thus higher prevalence of age-related hearing loss, improved early detection and diagnosis of hearing impairment, increased use of ototoxic medications (e.g., in the treatment of neonatal infections, malaria, drug-resistant tuberculosis, HIV, or cancer), and growing urbanization in many countries and exposure to potentially harmful levels of environmental and occupational noise (Olusanya et al. 2014). Recent efforts by the WHO have focused attention on strategies for reducing the growing burden of hearing disorders worldwide and have established guidelines for primary and secondary prevention of disabling hearing impairment (WHO 2006). Research that will improve the understanding of the epidemiology of hearing impairment worldwide can inform public health policy and is critical to the development of effective preventive interventions and should therefore be a global health priority.

Conflict of interest Gary Curhan declares that he has no conflict of interest.
Sharon Curhan declares that she has no conflict of interest.

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Chapter 3
Hearing Aid Transducers

Mead C. Killion, Aart Van Halteren, Stefan Stenfelt, and Daniel M. Warren

Abstract This chapter contains a brief historical and descriptive review of the microphones, earphones, and bone vibrators that are the essential elements in a hearing aid. The dramatic reduction in size of microphones and earphones (receivers) is documented, as is their improved performance with time. A discussion of their *theoretical* performance (sensitivity, noise, and output) versus size is followed by a comparison of theory and practice. The practical effects of microphone location about the ear and eartip location in the ear canal, and recent improvements in the ability to measure hearing aids, end the section on microphones and receivers. The final sections, on bone vibration history and progress, cover the progress to direct-to-bone vibrators.

Keywords Bone conduction receivers • Deep canal fittings • Directional microphone noise • Earphone receivers • Ear simulator • Frequency response • Microphone distortion • Open canal fittings • Percutaneous • Real-ear response • Subcutaneous • Vibration • 0.4 cc coupler • 2 cc coupler

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Abbreviations

BA	Series magnetic microphone (Knowles)
BAHA	Bone-anchored hearing aid
CIC	Completely-in-the-canal
CMOS	Complementary metal oxide semiconductor
EK	Series electret microphone (Knowles)
ITC	In-the-canal
JFET	Junction field-effect transistor
KEMAR	Knowles electronic manikin for acoustic research
MEMS	Microelectrical mechanical systems
RECD	Real-ear coupler difference
SLM	Sound level meter
SPL	Sound pressure level
THD	Total harmonic distortion
WDRC	Wide dynamic range compression

3.1 Introduction—Historical Perspective

The primary transducers for a hearing aid minimally are a microphone and an earphone, often called a receiver. This chapter describes 100 years of hearing aid transducer progress. For much of that period, the most important dictum driving transducer development was "Make it smaller!" At the beginning of the twentieth century, the body-worn carbon-microphone microphones were larger than today's largest smartphones. By 50 years ago, hearing aid microphones and receivers were small enough to permit in-the-canal (ITC) hearing aids (which often lie mostly in the concha). By 25 years ago, transducers were small enough to permit completely-in-the-canal (CIC) hearing aids that can be essentially invisible under normal lighting. Figure 3.1 shows examples of hearing aid microphones used between the 1940s and now.

Progress has also been made in transducers and couplings used for delivering sound via bone conduction. Direct bone-anchored devices for hearing aids have resulted in a dramatic improvement in coupling efficiency, with increased maximum delivered output level and useful bandwidth. New devices designed to operate with intact skin offer hope for high-efficiency coupling but without skin-related problems.

Size was not the only issue 50 years ago. Feedback squeal from magnetic coupling between the magnetic microphone and magnetic receiver had only recently been solved by Carlson (1963), and feedback from vibration coupling had seen relief from the first vibration-canceling dual receiver introduced by Harada (1989).

A frequency response with annoying peaks and limited useful bandwidth was often blamed on transducer limitations, an undeserved stigma as demonstrated when the first high-fidelity hearing aid with 16-kHz bandwidth was described by Killion (1979), who demonstrated with extensive listening tests that the available bandwidth

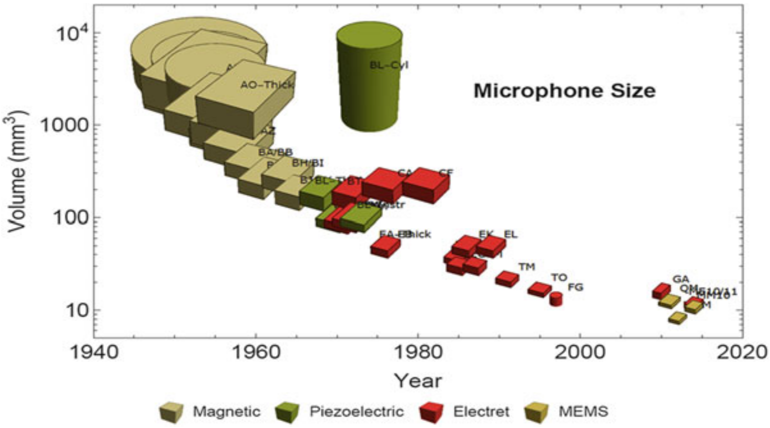

Fig. 3.1 Size of hearing aid microphones as a function of date of production since the 1940s (Graph courtesy of Knowles Electronics, LLC)

and response accuracy of hearing aid microphones and receivers were even then no longer a limitation. An experimental 16-kHz bandwidth hearing aid received higher fidelity ratings than the popular monitor loudspeakers used in Chicago professional music recording studios at the time. Later on, the limited bandwidth of early digital hearing aids was often blamed on the transducers, even though the digital-sampling clock had already made a bandwidth above 5–6 kHz impossible and the use of excessively short digital recovery times further reduced fidelity.

For many users, the primary complaint with hearing aids was that they did not solve the most pressing problem; users still could not hear well in noise. The problem was made worse by limited bandwidth and peaks that induced the listener to turn down the average gain.

Digital noise reduction was offered as a solution to the problem of hearing in noise, but despite the suggestion of some early ads, digital noise reduction has never produced an improved intelligibility in noise when the target is speech and the noise is unwanted speech or speech-shaped noise (Ricketts and Hornsby 2005; Nordrum et al. 2006; Bentler et al. 2008; Pittmann 2011). Thus in the future, as in the past, it appears the only improvements in hearing in noise must come from transducers designed to reduce noise before entering the hearing aid, using directional microphones, array microphones, and remote microphones.

On the positive side, the intrinsic noise level of today's hearing aid microphones is no longer a problem, regardless of technology. Experimentally, subjects wearing hearing aids with much more gain than would be used in normal practice have been able to detect soft sounds nearly as well as subjects with normal hearing (Killion 1992). Technically, aided sound-field thresholds close to 0 dB HL have been obtained. A recent "quad-element" microelectrical mechanical systems (MEMS) microphone for hearing aids competes nicely for noise level with electret microphones, and both types are available in 1 mm thickness and 10 mm^3 total volume.

Modern hearing aid transducers have virtually eliminated the bandwidth and response limitations of the past. Microphones and receivers with 16-kHz bandwidth and high-fidelity response are now available, which is more than twice the bandwidth of many digital hearing aid circuits.

As described in this chapter, the transducer size problem has also been largely solved. At the moment, from one or the other of the major transducer manufacturers, a 1.07-mm-thick electret microphone, a 1-mm-thick MEMS microphone, and a 1-mm-thick balanced armature receiver are available. Although not a practical consideration, it is interesting to note that those receivers are small enough so that 11 of them can be stacked inside a 7.1-mm-diameter circle, roughly the average diameter of the ear canal near the eardrum.

3.2 Transducer Types

3.2.1 Microphones

The first hearing aid microphones were carbon microphones borrowed from telephones. The first head-worn hearing aid microphones were magnetic. They became practical with the availability of transistor amplifiers, which, however, had a relatively low-input impedance of typically 5 kΩ. Thus good energy efficiency was required from the microphone to provide adequate sensitivity when the microphone was connected to such a low-input impedance. The few picofarads capacitance of a typical electret-condenser microphone would have been effectively shorted out. By one estimate, the resulting noise level would have been equivalent to 75 dB sound pressure level (SPL), completely masking normal conversational speech.

Once low-noise junction field-effect transistor (JFET) preamplifiers became available, however, energy efficiency in the microphone ceased to be important, and the higher voltage sensitivity of piezoelectric ceramic (Killion and Carlson 1974) and electret microphones made them far superior to magnetic microphones. Similarly, MEMS microphones—described in some detail in Sect. 3.2.1.2—are practical today even though a single-element MEMS microphone may have an active capacitance of less than 1 pF.

3.2.1.1 Electret Microphones

Once subminiature electret microphones were introduced nearly 50 years ago, they quickly replaced magnetic and ceramic microphones. The new microphones had lower noise, more than twice the bandwidth, and a dramatically reduced vibration sensitivity. Figure 3.2 shows an early electret microphone using a "diaphragm on bumps" construction. The electret material is charged to a high voltage and the "loose charges" are removed by conditioning, leaving several hundred volts of stable charge remaining.

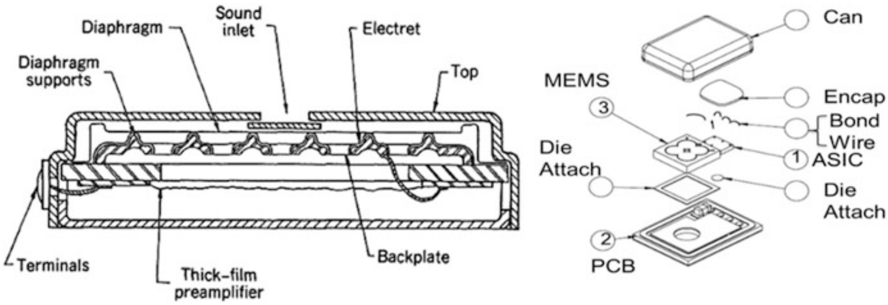

Fig. 3.2 Microphone constructions. (*left*) Electret-condenser microphone with built-in preamplifier. (*right*) MEMS microphones (Graph courtesy of Knowles Electronics, LLC)

Although the electret microphone construction shown in Fig. 3.2 provided a sensitivity essentially independent of temperature and humidity, subsequent electret microphone designs intended for hearing aids have traded stability for significantly lower noise performance and exhibit sensitivity versus humidity coefficients of 0.02–0.06 dB/% RH. This not a practical problem for omnidirectional hearing aid applications because humidity extremes seldom occur in one climate and wide dynamic range (WDRC) compression typically reduces a 4-dB change to a 2-dB change, about the minimum detectable change and not likely to be noticed in practice. The large effect of humidity can be much more of a problem in twin-microphone directional hearing aid applications, where a 1-dB change in sensitivity *between* the two microphones can degrade the performance to nearly omnidirectional at low frequencies.

3.2.1.2 MEMS Microphones

The ability to micromachine doped silicon wafers to precision dimensions produced a cost and uniformity that helped them become the microphone of choice for cellphones: some 2 billion were produced in 2014. For many years the application of MEMS microphones to hearing aids was limited by their noise levels. In 2011, a "quad" MEMS microphone with 25 dBA SPL equivalent of the electrical noise level (Warren 2011), 28 μA battery drain, and a flat frequency response up to 20 kHz was introduced, with dimensions of $1 \times 2.5 \times 3.35$ mm (less than 9 mm^3 in volume).

Both electret and MEMS microphones are moving-diaphragm condenser microphones. Sound pressure changes cause changes in the capacitance between the diaphragm and a charged surface. In an electret microphone, the charge is supplied by an electret material, while in a MEMS microphone, it is supplied by a charge pump when power is applied to the microphone.

With the same processes used to fabricate integrated circuits, successive deposition, masking, and etching occur on a wafer substrate to build up the layers of the MEMS microphone, as shown in Fig. 3.2. Once a wafer is fabricated, it is cut into individual dice, and after finishing, a functioning transducer results. A specially designed complementary metal oxide semiconductor (CMOS) preamplifier is added to form the completed microphone (Chang 2006).

One advantage of MEMS design is a stable sensitivity that is virtually unchanged as temperature and humidity change. This improved stability is possible because all the components have the same expansion characteristics.

Another significant advantage of MEMS technology is cost. There can be tens of thousands of microphones on a single wafer, whereas electret microphones like those used in hearing aids are typically produced one at a time. MEMS construction also allows submicron control over the most critical dimension of the microphone with regard to performance. Improved matching—and stability of matching—of frequency responses of two microphones in dual-microphone directional systems can result in improved directional performance and one that drifts little over time.

3.2.2 Receivers

The miniaturization of receivers has kept pace with that of microphones, as illustrated in Fig. 3.3, which shows changes in subminiature magnet receivers for head-worn hearing aids over a 60-year period. (The first author was design leader on the receiver shown in Fig. 3.4a, which is still in production today, nearly 50 years later; Knowles Electronics, 1965.)

The most energy-efficient receiver designs are magnetic, typically using a push–pull type of "balanced-armature" magnetic construction such as illustrated in Fig. 3.4.

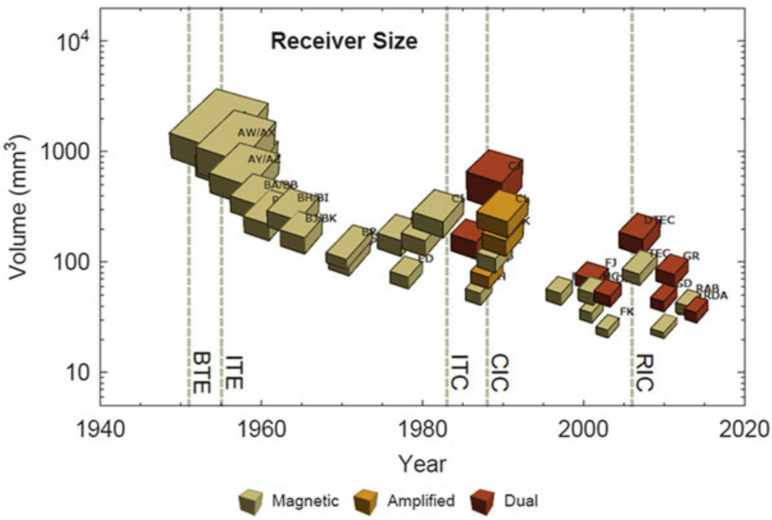

Fig. 3.3 Size of hearing aid receivers as a function of date of production since the 1940s (Graph courtesy of Knowles Electronics, LLC)

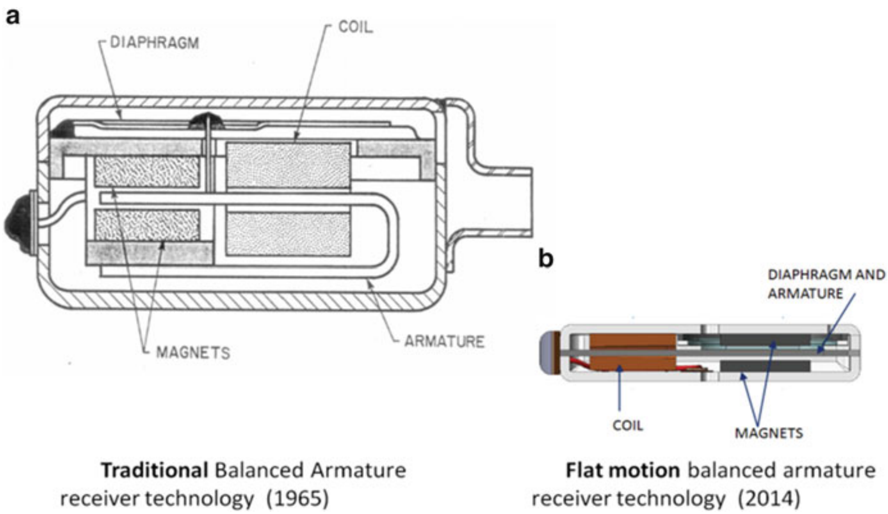

Fig. 3.4 Two balanced-armature receivers. (**a**) Traditional construction, which uses a separate armature and diaphragm. (**b**) Flat motion construction that combines the armature and diaphragm, permitting a total thickness of 1 mm (Drawings courtesy of Sonion Group and Knowles Electronics, LLC)

Other transducer mechanisms have been proposed, but to the writers' knowledge none come close to providing the same electromechanical coupling coefficient. Piezoelectric bimorphs, the next nearest competitor, are a factor of 10 less efficient in converting electrical energy to acoustic energy. The electromechanical coupling coefficient for balanced-armature receivers is $k=0.8$, while for bimorphs $k=0.25$, so the proportion of electrical energy converted to the mechanical side (k^2) is 0.64 and 0.06, respectively. The dominance of the magnetic receiver is thus readily understood. Hearing aid wearers must carry the power source around with them, and the receiver often consumes half or more of the power in a hearing aid. Thus, receiver efficiency sometimes determines almost directly how often the battery must be changed.

3.3 Transducer Performance Versus Size

The bandwidth and frequency response smoothness of both microphones and earphones have steadily improved with their decreasing size, as has nearly every other property, including resistance to shock damage, magnetic shielding, and microphone insensitivity to vibration.

The minimal shock mounting routinely used in ITE and canal hearing aids has been made practical because of increasingly rugged transducers. One modern hearing aid receiver will withstand 20,000 g shock, produced, for example, by a drop from 2 m with a stopping distance of 0.1 mm. Modern microphones will all withstand 20,000 g or more without damage.

It is perhaps not surprising that the bandwidth of receivers has generally improved with smaller size. Miniature magnetic receivers have a real-ear frequency response that is intrinsically flat from as low a frequency as you would wish (less than 1 Hz with a high-quality earmold seal and a minimal-size barometric-release vent inside the receiver) up to the frequency where the combined acoustic-mechanical masses and compliances resonate. Because armature compliance increases with the cube of length, all other things being equal, while armature mass increases only linearly with length, smaller mechanical devices have a natural tendency toward higher resonance frequencies.

3.3.1 Omnimicrophones

3.3.1.1 Frequency Response

The bandwidth of electret microphones is restricted primarily by the amount of acoustic mass (inertance) in the inlet sound channel. A long, small-diameter inlet channel can reduce the bandwidth to 4 kHz. In contrast, hearing aid electret microphones that provide free access to the diaphragm with a "salt shaker" arrangement of many holes in the cover have offered a flat frequency response with bandwidths of 16 kHz since the late 1960s: they have been used in broadcast and recording studies for more than 40 years. The most recently available MEMS microphones offer nearly flat response to 40 kHz.

3.3.1.2 Maximum Undistorted Input SPL

In most cases, the maximum sound level the hearing aid can handle without distortion is limited by the input analog-to-digital (A/D) converters; see Whitmer, Wright-Whyte, Holman, and Akeroyd, Chap. 10. The peak–peak voltage swing for a hearing aid microphone preamplifier is typically 600–900 mV before clipping, corresponding to about 200 mV_{rms}. (In this chapter, all numbers are rounded for simplicity.) If the microphone has a sensitivity of −36 dBV/Pa, that is, 16 mV_{rms} at 94 dB SPL, then the 200-mV_{rms} microphone output corresponds to 116 dB SPL for a sine wave or 119 dB SPL for the instantaneous peaks (200 mV is 22 dB above 16 mV).

This high peak-handling capability is required for listening to music. Peak sound level meter readings of 104 dB(C) SPL have been measured at several Chicago Symphony Orchestra concerts in the first balcony. A reading of 104 dB SPL on a sound level meter (SLM) corresponds to an instantaneous peak of 114–116 dB SPL. Even with the higher sensitivity of −33 dBV/Pa of some microphones, the microphone itself will almost never overload, even at a symphony or jazz concert (as opposed to an amplified rock concert).

Working forward into the hearing aid circuit, to preserve the capability of typical hearing aid microphones, the input stage of the hearing aid and the A/D converter must be able to handle 600–900 mV P–P without clipping. Most digital hearing aids don't have this much "headroom" according to the data presented in Table 3.1,

Table 3.1 Maximum input level of hearing aids is defined by a rapid rise in 3rd harmonic distortion, corresponding to clipping

Max. on 312	Hearing aid A	Hearing aid B	Hearing aid C	Hearing aid D	Hearing aid E	Hearing aid F
Input handling	Input AGC	Input AGC	Input clipping	–	Input clipping	Input clipping
Max. input	92 dB SPL	95 dB SPL	102 dB SPL	108 dB SPL	103 dB SPL	113 dB SPL
Sampling frequency (max bandwidth)	– (10 kHz)	– (7.9 kHz)	– (7.6 kHz)	– (7.6 kHz)	33.1 kHz (10.4 kHz)	33.1 kHz (11 kHz)
Bit depth	–	–	–	–	16 bit	16 bit

From Jessen (2013)

which may help explain why some hearing aids annoyingly distort at live classical and jazz music concerts; see Chap. 8. There is good news, however. The bandwidths in Table 3.1 are generally improved from those of 10 years ago.

3.3.1.3 Noise

All things being equal, the equivalent SPL of microphone noise for any microphone that is small compared to a wavelength should be roughly proportional to the square root of the microphone volume, so twice the volume should reduce the noise level by 3 dB. This can be seen if two microphones are placed close together and their electrical outputs are connected in series, assuming separate batteries for this thought experiment. The resulting *signal* sensitivity will be 6 dB greater because the signal voltage outputs of the two microphones are equal and coherent (in phase) and thus add linearly. Their combined *noise* output will be only 3 dB greater, however, because their individual noise voltage outputs will be generally completely uncorrelated (random) and thus add in power. In practice, microphones are normally connected in parallel, but the result is the same. In this case, the signal voltage is unchanged but each microphone loads the other, so the individual noise level from each is 6 dB lower and the combined noise is 3 dB lower.

Although this reasoning holds in general, improvements in materials and design can sometimes provide surprising results. As an example, based on size considerations, the Knowles EK series electret microphone might be expected to have a 9 dB higher equivalent noise than the earlier Knowles BA-series magnetic microphone because the BA microphone has eight times greater volume; in fact, the EK microphone is just as quiet. While the energy efficiency of the EK microphone is much lower than that of the BA, the EK's intrinsic noise level is lower still because of its lower internal losses (lower real part of its acoustical, mechanical, and electrical impedances).

The equivalent input level of the noise in a 1-Hz bandwidth at 1 kHz is about −14 dB SPL for both microphones. At 2 kHz and above, the effect of the head and pinna increases the SPL at the microphone inlet of an ITE hearing aid by approximately 5 dB,

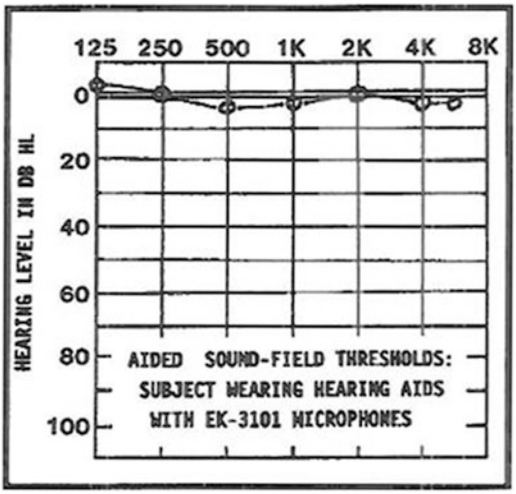

Fig. 3.5 Illustration of the fact that, with sufficient gain in the hearing aid, aided thresholds approximating 0 dB HTL can be obtained with a microphone having an A-weighted noise level equivalent to 25 dB SPL (Used with permission from Killion 1992)

which acts to decrease the apparent microphone noise level by a comparable amount. As a check on this theoretical prediction, normal-hearing subjects were tested with enough gain (approximately 10 dB) so that the microphone noise, rather than their own thresholds, would dominate. As shown in Fig. 3.5, thresholds between 0 and 5 dB HL across frequencies were obtained (Killion 1992). Although only a hearing aid wearer with normal low-frequency hearing might be expected to notice the corresponding microphone noise levels, hearing aids with microphones having 10 dB higher noise level have generated complaints in the field.

Figure 3.6 shows the relationship between noise and volume over the years. By way of comparison, the 640AA 1″ condenser microphone developed in the 1940s and used in the Bell Labs dummy "OSCAR" has an equivalent A-weighted noise level of 13 dB SPL, 7 dB below the apparent noise of the ear (Killion 1976). With its vacuum preamplifier, this microphone occupied 50,000 mm^3. A 1976 experimental hearing aid microphone had an equivalent noise level of 20 dB SPL in a 200 mm^3 volume (Killion 1976). It is now possible to obtain an equivalent noise level of 16.5 dB SPL in a 100 mm^3 volume by electrically paralleling seven 2.5-mm-diameter microphones, resulting in a 7.5-mm-diameter assembly (see insert in Fig. 3.6).

3.3.1.4 Vibration Sensitivity

The acoustic gains routinely obtained in modern ITE and canal hearing aids, with the microphone and receiver almost touching each other, are a direct result of reduced magnetic and vibration coupling between microphone and receiver. The remaining limitation is acoustic coupling in the form of sound leakage from the ear canal back to the microphone opening and the SPL generated on the face of the

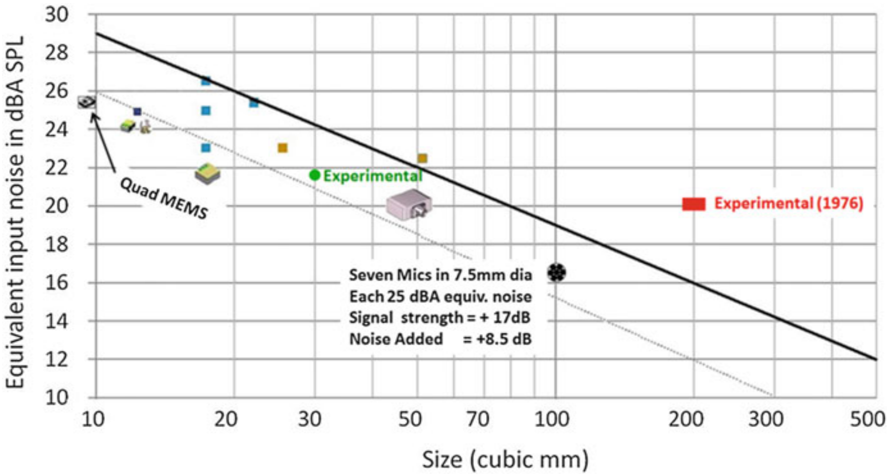

Fig. 3.6 Equivalent input level of A-weighted microphone noise versus volume. (Graph from Sonion Group and with combined data from Knowles Electronics, LLC, and from Killion. Note the inset comment)

hearing aid by receiver vibration (see Sect. 3.3.3.1). This leakage can be due to intentional "venting" of the earmold (including "open" fittings) or an inadequate seal of the earmold in the ear canal.

The *direct* mechanical coupling to the microphone produces little feedback problem because the mechanical vibration sensitivity of most recent electret microphone designs is so low it is nearly impossible to measure (Killion 1975). The most recent microphones can be safely cemented directly to the hearing aid housing with either no increase in feedback problems or, when previously a rubber coupling tube allowed the microphone to "pump" on the tube, a decrease in feedback problems.

Over the last 50 years, the vibration sensitivity of hearing aid microphones has dropped from approximately 106 dB SPL equivalent for one g of vibration (1960s magnetic microphones) to 74 dB SPL (1970s electret microphones) to less than 60 dB SPL (latest electret microphones).

3.3.2 *Directional Microphones (Single Cartridge and Dual Microphone)*

3.3.2.1 Effect of Location in Ear

In addition to the standard omnidirectional electret microphones, directional microphone capsules with a variety of internal rear-port time delays make it possible for a hearing aid manufacturer to produce different directional characteristics with a given port spacing or to accommodate different port spacings. A properly utilized directional microphone provides an improvement of 3–5 dB in signal-to-noise ratio even under difficult reverberant listening conditions and can provide a much greater benefit in the open. Killion et al. (1998) reported a 9–12 dB benefit on Bourbon Street in New Orleans.

To obtain good in situ performance in directional microphone hearing aids, the effect of head and ear diffraction must be taken into account. The location of the microphone has a marked effect on the effective port spacing. An over-the-ear location increases the effective spacing to about 1.4 times the physical spacing, whereas an ITE location decreases the effective spacing to about 0.7 times the physical spacing. Note: in free space for frontal sound, the time delay between inlets is 3 μs for each 1-mm spacing. For a cardioid characteristic with a 10-mm spacing, therefore, the internal time delay must be 30 μs to cancel the external delay.

Many early directional microphone hearing aid designs had poor directivity for frequencies above 1–2 kHz, but Madaffari (1983) demonstrated that with careful attention to time delays, inlet and outlet phase shifts, and case and head diffraction, it is possible to make directional microphone hearing aids whose in situ directivity is good from low frequencies up to 4 or 5 kHz. Because of the lack of pinna shielding, the directional microphones in behind-the-ear hearing aids typically have 1.5 dB lower directivity than the same microphones in an ITE hearing aid. Figure 3.7 illustrates the directional characteristics obtained with BTE and ITE microphone locations, as well as those of an array microphone mounted over the ear.

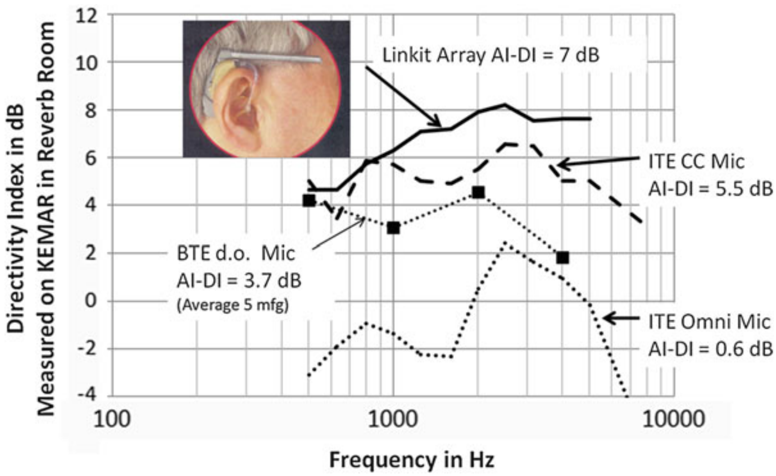

Fig. 3.7 Directivity of Omni, typical BTE dual-omni (d.o.), ITE, and Array hearing aid microphones (BTE d.o. data courtesy Starkey Inc.; the remaining data from Etymotic Research reverberation room)

3.3.2.2 Effect of Microphone Mismatch in Dual-Microphone Designs

Although early directional microphones for hearing aids used a single-cartridge construction with internal acoustic resistor-capacitor time delay, more recently dual-omnidirectional microphones have been used, with the time delay supplied in the digital processor. One of the problems with dual-microphone directional constructions is the sensitivity to microphone mismatch. Differences of a few hertz in the "corner frequency" of roll-off between the two microphones can cause the resulting polar plot to point backward at low frequencies (Warren, personal communication, 2014). The unit-to-unit differences in corner frequencies are substantially lower in MEMS microphones, reducing the likelihood of a dual-omnidirectional microphone pointing backward at low frequencies. Note: a microphone mismatch at 125 Hz of 0.05 dB and 0.8° can change a hypercardioid into a backward bulging figure-eight directional response (Warren, personal communication, 2014).

3.3.2.3 Array Microphones

Soede et al. (1993) developed a four-microphone array that was later improved by Etymotic Research with his help to provide an 8-dB directivity at high frequencies, as shown in Fig. 3.7. It was an "additive" design so the noise performance was good. Unfortunately, the enthusiastic reception to this greater directivity was limited to relatively few hearing aid wearers. For the rest, the cosmetic/ergonomic features appeared to outweigh the benefit.

One array microphone that was successful was the Siemens TRIANO model, which used three microphones to form a beam (Powers and Hamacher 2002, 2004). It was a "subtractive" design, so the inherent frequency response fell at 12 dB per octave at low frequencies and the gain needed to restore the low-frequency response would have made the internal noise unacceptable at low frequencies. As a compromise, first-order directivity at low frequencies was combined with second-order directivity at high frequencies, resulting in overall high-frequency directivity of 8 dB, as measured on the KEMAR manikin.

3.3.3 Receivers

3.3.3.1 Receiver Sensitivity Versus Size

The dominant impedance determining the motion of a subminiature receiver's diaphragm is not that of the 0.6 cc or so of air in the occluded ear canal or of the total 1.3-cc equivalent volume when the compliance of the eardrum is included but that of the much smaller volume of air trapped *behind* the receiver diaphragm, which is less than 0.04 cc in the smaller receivers. The diaphragm produces the same volume displacement into the back volume as it does in the ear, so that producing 115 dB SPL in the ear canal at low frequencies requires the production of 145 dB SPL in the "back volume" behind the receiver diaphragm.

Although the use of two receivers produces nearly a 3-dB improvement in undistorted output level for the same electrical drive, the more important advantage has been vibration cancellation. As described by Harada (1989), if the two halves are driven electrically in phase but mounted so their mass reactions are opposite in phase, a substantial reduction in vibration-induced feedback results.

So, with both microphones and receivers, halving the size theoretically costs 3 dB in performance, all other things being equal. As receivers have decreased in size dramatically—11 of the smallest one could be stacked in the diameter of an ear canal—the size reduction has come at the expense of the 3 dB/doubling theory: the smallest microphones have less noise than predicted from their size reduction, while the smallest receivers have less output than predicted from their size reduction.

3.3.3.2 Frequency Response

It is perhaps not surprising that the bandwidth of receivers has generally improved with decreasing size, as discussed in Sect. 3.3.3.1. It has been known since the 1970s that the real-ear response of subminiature receivers could match the open-ear response of the ear in a diffuse sound field (Killion and Tillman 1982). It is also possible to produce a flat pressure response for frequencies up to 16 kHz, as shown in Fig. 3.8.

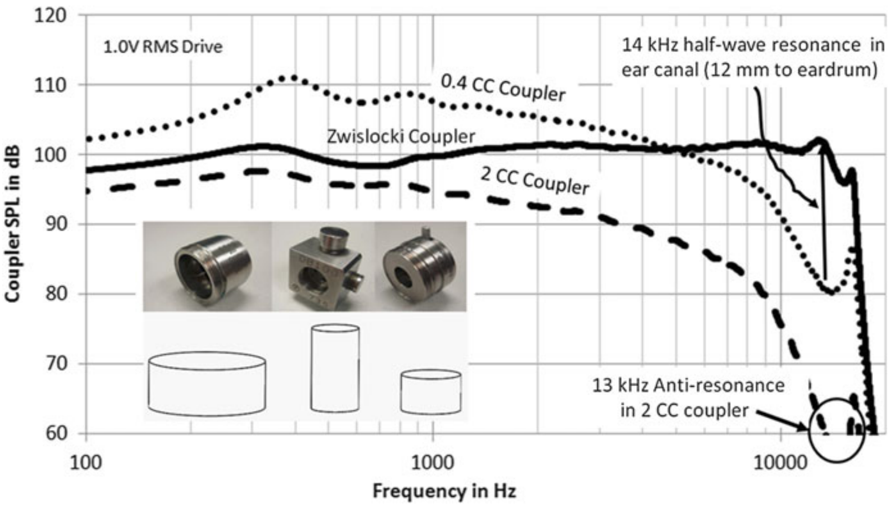

Fig. 3.8 Comparison of frequency response of ER-2 insert earphone in a 2-cc coupler, Zwislocki coupler ear simulator (similar to "711" ear simulator), and a 0.4-cc coupler

3.3.3.3 Allowable Distortion

While total harmonic distortion (THD) of a few tenths of 1 % can be detected in a single-frequency tone under carefully contrived experimental conditions, the just-detectable distortion for speech and music is somewhat higher. Near threshold, the harmonics produced by distortion may be simply inaudible. At high levels, the apparent distortion in the ear masks the harmonics. A 2 % THD is generally inaudible for speech and music between 50 and 90 dB SPL. Below 50 dB SPL the inaudible distortion rises to 10 % at 30 dB SPL. Above 90 dB SPL, the inaudible distortion rises linearly to 10 % at 110 dB SPL. The derivation of the limits can be found in Killion (1979).

3.3.3.4 Maximum Output for Various Size Receivers

Over the past 50 years, the size of subminiature balanced-armature receivers has fallen from 350 mm^3 to 14 mm^3. Although the theoretical expectation is that the peak output should fall 3 dB for each halving in volume, in practice the relationship over the last six generations of hearing aid receivers has been closer to 5 dB each time the volume is halved. The minimum thickness of a magnet required to produce a given field does not scale, for example, so that a disproportional drop in output is expected as the thickness of the receiver decreases.

The largest (dual) receivers can produce a peak output level in a 2-cc coupler of 143 dB SPL; the smallest can produce 117 dB SPL. Many of the larger receivers are still in use in high-power hearing aids.

3.3.4 Receiver Real-Ear and Coupler Measurements

Unlike microphones, whose response can be readily measured in the free field, the outputs of hearing aid receivers must be measured in some type of coupler that more or less resembles the human ear (ANSI 1973). The 2-cc coupler introduced by Romanov (1942) was an easily reproduced device designed to minimize cross- and circumferential modes for frequencies up to about 8 kHz, but it fails to reproduce the acoustic properties of a human ear and has a deep null around 13 kHz (see Fig. 3.8).

Zwislocki (1970) described a real-ear simulator that was later modified slightly by Burkhard and Sachs (1977) to agree with their probe-microphone measurements on 11 real ears. Subsequently, the European “711” coupler gave nearly identical performance (ANSI 1979). When used with the KEMAR manikin (Burkhard and Sachs 1975), both permitted, for the first time, frequency-response measurements and sound recordings that were equivalent to those made in an average human ear with average pinna, as well as including the diffraction of the head and body. The Zwislocki coupler ear simulator also permitted accurate measurements of the effect of the depth of the seal of the eartip and/or the depth of the sound outlet in vented or “tube” fittings; more recent ear simulators do not.

Ear simulators are expensive and are susceptible to response changes from contamination by debris that may not be visible. This led to several attempts over the years to devise a simple small-volume coupler that would provide response measurements for frequencies up to 16 kHz and that could be reliably related to the response in the real ear.

3.3.4.1 The 0.4-cc Coupler

There is evidence from several laboratories suggesting that a 0.4-cc coupler can provide a real-ear coupler difference (RECD) and, what is equivalent, an ear-simulator-coupler difference (Frye 1995), which is similar across various hearing aid sound sources, ranging from traditional BTE aids with No. 13 tubing and sealed earmolds to receiver in the canal (RIC) aids (Gebert and Saltykov 2013). Aldous and Stewart (2014) provide a reasonable summary of the supporting experimental data from Etymotic Research, obtained with the 0.4-cc coupler described by Gebert and Saltykov (2011). Later measurements showed identical results with a G.R.A.S. 0.4-cc coupler (2013).

Figure 3.8 shows the frequency response of an Etymotic ER-2 insert earphone measured in three couplers. Table 3.2 reports the volume of each coupler. In the case of the Zwislocki coupler, only the canal portion of its volume is shown; the equivalent volume of the eardrum simulation is not included.

3.3.4.2 Universal Real-Ear to 0.4-cc Coupler Difference

Aldous and Stewart (2014) also investigated the effect of acoustic source impedance on RECD curves using four sound sources ranging from low impedance (large CI series receiver through BTE coupling) to high impedance (small ED series receiver with damped 12-mm tubing) to very high impedance (EH receiver with 75 mm of 1-mm internal diameter tubing). The resulting RECD curves are shown in Fig. 3.9. Fortunately, the RECD curves are sufficiently similar to suggest that the 0.4-cc coupler was an acceptable solution to the problem of a simple, easily produced wide-band coupler, whose results could be used to predict real-ear results for sealed earmolds (which is how hearing aids are normally tested). Three companies now offer a 0.4-cc coupler.

Table 3.2 Volume of each coupler shown in Fig. 3.8

Coupler	Diameter in mm	Length in mm	Volume in cc	Diameter in cm	Length in cm
2 cc	18.5	7.44	2	1.85	0.744
Zwislocki	7.5	13.0	0.575	0.75	1.30
0.4 cc	9.45	5.70	0.4	0.945	0.570

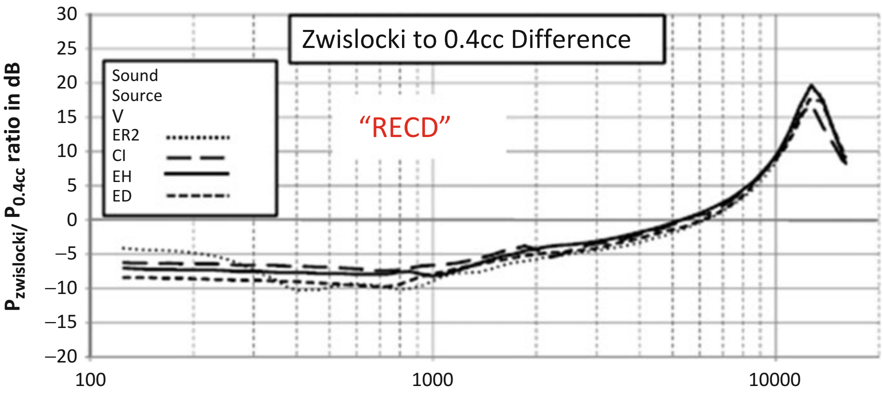

Fig. 3.9 Effect of acoustic source impedance on the "RECD" for Zwislocki to 0.4-cc coupler. Four sources are shown: damped ED (ER-2 with 300-mm tubing); damped ED with 12-mm coupling; undamped EH with 75 mm × 1 mm tube; undamped CI with 75-mm total length BTE coupling

3.3.4.3 Real-Ear Effects on RECD

Eartip depth and degree of venting (including tube fittings) can have a much larger effect on the real-ear response than the small differences shown in Fig. 3.9.

Normal Closed-Canal Fittings

For standard closed-canal fitting, the ear does not present a simple volume compliance. With an eartip seal halfway down the ear canal, the residual ear canal volume is 0.7 cc, to which is added the equivalent volume of the eardrum, about 0.65 cc at low frequencies (Jerger et al. 1972), with a resulting total equivalent volume of 1.35 cc. At low frequencies, this produces 3.5 dB greater sound level at the eardrum than in a 2-cc coupler when using a typical hearing aid receiver, whose volume is about 0.05 cc. At midfrequencies, the motion of the normal eardrum appears as an acoustic resistance of approximately 300 cgs acoustic ohms, reflected back through the middle ear from the cochlea. Above several kilohertz, the mass of the eardrum and ossicles dominates the compliance, and there is only a small movement of the eardrum compared to the low-frequency case. In this case, the "load" is essentially the 0.7-cc residual volume of the ear canal (Goode et al. 1994). The result of all of these factors is the well-known RECD curve for the 2-cc coupler, based on the data of Burkhard and Sachs (1977), which shows the easily remembered values of 3.5 dB at low frequencies, 5 dB at 1 kHz, 10 dB at 3 kHz, and 15 dB at 6 kHz. (Those same differences can be seen in Fig. 3.8.)

Deep-Canal Fittings

Two advantages result from deeper fittings. First, there is a reduction in the occlusion (hollow voice) effect (Killion et al. 1988), and second, a greater undistorted output can be obtained from a given size receiver. When the eartip is moved closer to the eardrum, the RECD increases more at high than at low frequencies. With a deeply sealed eartip, the RECD becomes 7 dB at low frequencies, 9 dB at 1 kHz, 16 dB at 3 kHz, and 22 dB at 6 kHz.

By using a deeply inserted sealed eartip, a RIC hearing aid with only 96 dB SPL output in a 2-cc coupler at 4 kHz can be expected to produce 115 dB SPL at the eardrum. The earliest CIC hearing aids and the Lyric deeply inserted hearing aid benefit from this increase.

Open-Canal or "Tube" Fittings

The response of an open-canal fitting depends somewhat on the depth of the sound outlet, as illustrated in Fig. 3.10. Surprisingly, a drop in eardrum pressure and rise in feedback pressure occurs with deep tube fittings. This comes about because at 5.6 kHz the half-wave resonance of the open ear acts to cause a dip in pressure developed at the eardrum (at half-wave resonance, a tube tends to "acoustically disappear" as a back pressure) while the same resonance causes a large peak in the sound coming *out* of the ear. A location only one-third of the way down the ear

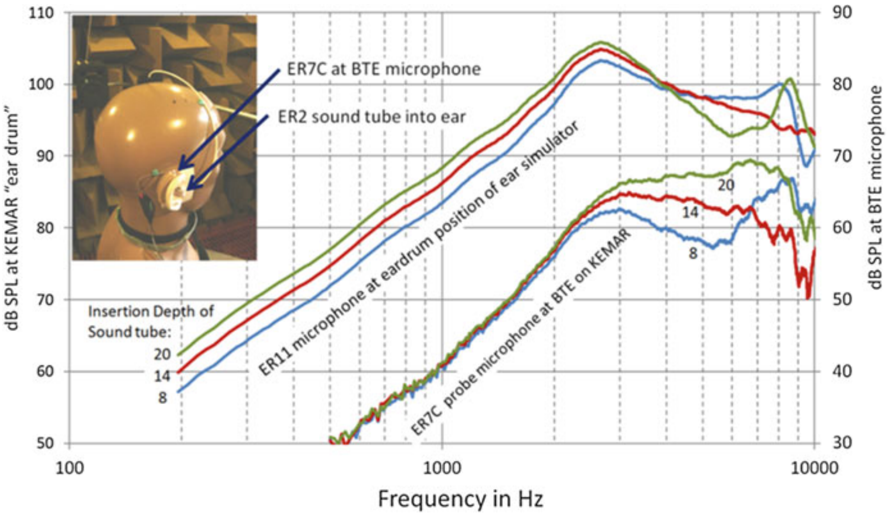

Fig. 3.10 Vent response and feedback sound level for open-ear or "tube" fittings. At 5 kHz, a deep insertion results in 12 dB lower gain before feedback, depending on the phase between the eardrum pressure and microphone

canal can provide 12 dB more gain before feedback than a deeply fitted tube. Fortunately, digital feedback reduction methods have reduced sensitivity to feedback in open-canal fittings, although sometimes at the cost of unusual “tweets” when pure tones are presented.

3.3.5 Bone Conduction Receivers

Bone conduction refers to sound transmitted through the skull bone and soft tissue that results in a traveling wave on the basilar membrane, leading to auditory perception (Stenfelt and Goode 2005b; Stenfelt 2011). When a bone conduction transducer is coupled to the skull bone directly or via the skin covering the skull bone, the vibration is transmitted to both cochleae (Stenfelt and Goode 2005a). Bone conduction can also be produced from airborne sound, but for the normal ear the air conduction route has 40–60 dB better sensitivity than the bone conduction route (Reinfeldt et al. 2007).

One of the first discoveries related to bone conduction was that a vibrating rod touching the teeth became audible (Berger 1976). This sparked the idea that sound could be transmitted to the ear by means of vibrations applied to the teeth or to the skull, for example, at the forehead or the mastoid. This led to the development of the first passive bone conduction hearing devices for people with conductive hearing losses. These devices used a rod of wood or metal to transmit vibrations either to the teeth or to the skull. One device had a speaker end of the rod that was pressed against the talker’s larynx while the other end of the rod was coupled to the teeth or mastoid of the listener (Berger 1976). Another common type of device, termed the Dentaphone, used a large thin area (diaphragm) to pick up sound vibrations in the air and coupled these to a rod placed against the teeth (Berger 1976).

The number of bone conduction hearing aids used is a small fraction of the number of air conduction hearing aids. Therefore, the evolution of bone conduction hearing aids has relied to a large extent on the evolution of air conduction hearing aids. When the carbon microphone was introduced at the beginning of the twentieth century, the same technology was extended to bone conduction hearing aids. These consisted of an electromagnetic transducer pressed against the skin-covered mastoid behind the ear, held in place by a steel headband, and connected to a body-worn hearing aid processor. That basic design was kept with the use of spectacles instead of a headband, and ear-level hearing aid processors. To reduce problems with acoustic feedback, the processor with the microphone was placed on the opposite side of the head to the bone conduction transducer.

In the middle of the 1980s, devices that utilized osseointegration became commercially available (Mudry and Tjellström 2011). The most successful device was the bone-anchored hearing aid (BAHA) that used a skin-penetrating fixture to attach the device to the skull (Håkansson et al. 1985). The success of the BAHA led to commercialization of other devices that used bone conduction as the means of transmitting the sound to the inner ear (Popelka et al. 2010b; Hol et al. 2013), including implanting the whole transducer (Eeg-Olofsson et al. 2014; Manrique et al. 2014).

3.3.5.1 Load Impedance

One important factor that influences the function of the output transducer is the mechanical load impedance. For a bone conduction hearing aid, the load impedance is often that of the skull, but this is influenced by several factors. Of these, the most important are (1) whether the device rests on the skin or is directly attached to the bone, (2) location on the skull, (3) area of the attachment interface, and (4) for skin-applied transducers the static force of the attachment (Khanna et al. 1976; Stenfelt and Goode 2005a). Simplistically, the load from the skin-covered skull can be seen as a mass-spring-damper system where the spring is the compressibility of the skin and subcutaneous tissues, the mass is that of the skin and subcutaneous tissues that move with the transducer, and the damping comes from the skin and subcutaneous tissues (Flottorp and Solberg 1976; Stenfelt and Håkansson 1999). The typical mechanical impedance of the skin-covered mastoid is shown in Fig. 3.11. The mass of the head (3–4 kg) also influences the load impedance but only for frequencies below 40 Hz if the skin compliance value from Flottorp and Solberg (1976) is used. The spread of mechanical impedance values for humans causes the output from the bone conduction transducer to deviate by up to ±6 dB from the average value (Flottorp and Solberg 1976).

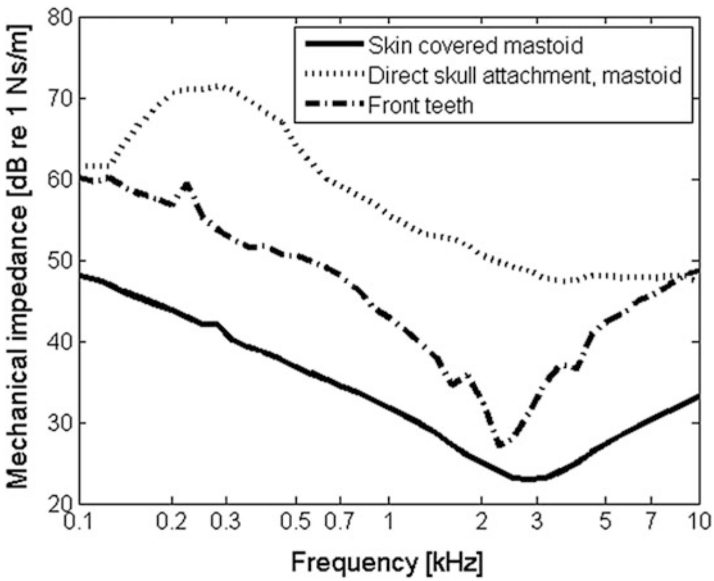

Fig. 3.11 Impedance magnitude curves for (1) skin-covered mastoid (*solid line*), (2) direct attachment to the skull bone at the mastoid (*dotted line*; data from Stenfelt and Goode 2005b), and (3) the front teeth (*dash-dotted line*; data from Stenfelt and Håkansson 1999)

Couplers with a representative mechanical load for measuring the characteristics of bone-conduction transducers have been developed for research (Haughton 1982; Stenfelt and Håkansson 1998) and commercially (e.g., Brüel & Kjær type 4930 artificial mastoid, Larson Davis AMC493B artificial mastoid). The mechanical impedance of such devices is specified in an IEC standard (IEC:60318-6, 2007) intended for the calibration of audiometer diagnostic bone conduction transducers. These devices must be used at an appropriate temperature to give correct results (Frank and Richter 1985). In general, all bone conduction transducer types need to be calibrated for their intended use. One way to accomplish that is to use a loudness balance method, where the loudness of sound evoked by the bone conduction transducer is adjusted to match that of sound in a calibrated sound field (Popelka et al. 2010a, b).

The load impedance differs when the soft tissue is eliminated and the transducer is attached directly to the skull, as for the BAHA system (Snik et al. 2005). As shown in Fig. 3.11, the impedance when attached directly to the skull is greater in magnitude and has less damping than the impedance of the skin-covered skull (Håkansson et al. 1986; Stenfelt and Goode 2005a). This means that the output of a bone conduction transducer is different when attached directly to the skull than when attached to the skin-covered bone. The magnitude of the mechanical impedance of the direct attachment to the skull peaks at frequencies between 150 and 400 Hz (Stenfelt and Goode 2005a). Below this frequency, the mass of the skull determines the impedance, whereas above this frequency the compliance of the bone surrounding the attachment point determines the impedance. For frequencies above 2–3 kHz, the local mass of the bone around the stimulation position also adds to the impedance.

It is necessary only for the coupler to have an impedance that is much greater than the output impedance of the bone conduction transducer to give a reliable estimate of the output force from the transducer. Therefore, the systems for measuring the characteristics of bone conduction transducers intended for direct skull bone attachment (Håkansson and Carlsson 1989) do not mimic the impedance of the skull but use the fact that the impedance of the skull is much greater than the output impedance of the bone conduction transducer. Such couplers do not estimate the output motion (displacement) of the bone conduction transducer correctly, which sometimes is an important measure of the transducer characteristic.

The third application point for bone-conducted sound is at the teeth (Dahlin et al. 1973; Stenfelt and Håkansson 1999; Popelka et al. 2010b). The impedance curve for the front teeth (incisors) in Fig. 3.11 resembles the impedance of the skin-covered skull but is overall higher. This curve might be different for other teeth, for example, the molars.

3.3.5.2 Electrodynamic Transducers

A bone conduction transducer transforms the electrical input to a mechanical vibration. That process can be accomplished by different mechanisms, often divided into four categories: (1) piezoelectric (electrostriction), (2) magnetostriction, (3) electrodynamic (moving coil and variable reluctance), and (4) electrostatic. The last of these requires high voltages and has so far not been used for bone conduction transducers. Magnetostrictive transducers have been reported for usage in bone conduction applications (Khanna et al. 1976; Popelka et al. 2010a) but are rare. The most common types of output transducers for bone conduction are electrodynamic, followed by piezoelectric, devices.

Moving coil devices (Fig. 3.12a) are commonly used in acoustic loudspeakers and exploit the force generated by the current in a coil in a magnetic field. The magnetic field is provided by permanent magnets designed to provide a homogeneous static magnetic field that surrounds the coil. The output force is then proportional to the current in the coil, leading to a linear output, theoretically without nonlinear distortion. Also, with proper selection of the masses, springs, and geometry of the moving coil transducer, a broadband response covering the audio frequency range can be achieved. However, these devices are often too large to be efficient, linear, and of broad frequency range. Consequently, they are not very commonly used for bone conduction transducers. However, the old KH70 audiometric bone transducer used this design and it could be used to test hearing for frequencies up to 16 kHz (Hallmo et al. 1991). Also, the upscale version of a floating mass transducer (Bonebridge) system uses this mechanism (Hassepass et al. 2015).

By far the most widely used design for bone conduction transducers is the variable-reluctance transducer (Fig. 3.12b). The widespread use is attributable to its simple design, based on a magnetic circuit with a gap, which is often air. The magnetic circuit tries to close the gap and thereby creates a force. If a permanent magnet is included in the magnetic circuit in series with a coil, the magnetic field at the gap can be modulated by the current in the coil and consequently generates a force. However, the force is not linearly related to the current and nonlinear distortion is created (Fig. 3.12c). The output force from this device depends on the mass of the transducer and the spring keeping the gap open, resulting in a resonant system.

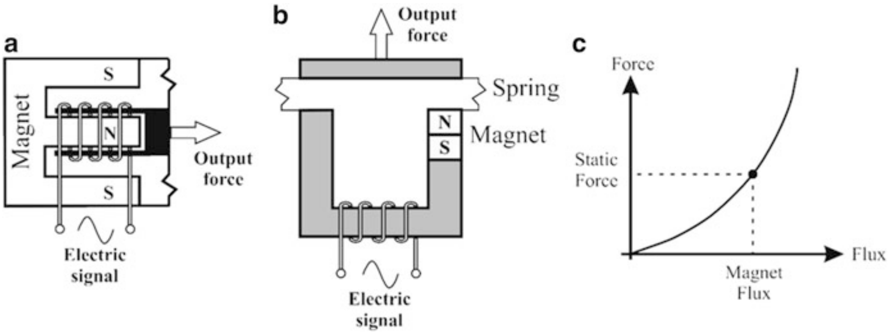

Fig. 3.12 (**a**) A schematic of a moving coil transducer. The magnet provides a constant magnetic flux over the movable coil. The output force is proportional to the magnetic field strength, the current in the coil, and the number of turns of the coil. (**b**) A schematic of the variable-reluctance transducer. The magnetic flux tries to close the air gap that is maintained by the springs. The electric signal modulates the flux and thereby the force in the air gap, resulting in a motion. (**c**) The output force from a variable-reluctance transducer is proportional to the square of the magnetic flux. The magnet biases the flux, resulting in a more linear response

This type of transducer has been used for the majority of bone conduction hearing aids in which the transducer is pressed against the skin. Also, the commonly used Radioear B71 diagnostic audiometric bone conduction transducer uses a variable-reluctance design. When attached to the skin-covered skull, the skin and subcutaneous tissue act as a damper and attenuate the peak of the transducer's resonance, so no inherent damping of the transducer is required. However, when attached directly to the bone, as with the BAHA, the sharp resonance of the transducer would limit the frequency range of the transducer and damping of this resonance is required. In the earlier versions of the transducer for the BAHA, damping material was inherent in the transducer design. However, the viscoelastic damping material used is strongly influenced by both temperature and humidity, and the transducer characteristics were thereby affected.

To overcome the problems associated with the damping material while avoiding the effects of the sharp resonance from the transducer, most bone conduction hearing aids in which the transducer is directly fitted to the skull bone use prefiltering of the signal. The filter is usually the inverse of the transducer's frequency response, resulting in a fairly flat frequency response. Normally, the transducer characteristics are stable over time and a stationary filter can be used, making it unnecessary to use adaptive filters. However, these devices are also used with a headband holding it on the skin, for example, when testing the device preimplantation or when surgery is not an option, such as with very young children. In such cases, the skin and subcutaneous tissue provide the necessary damping and if the prefiltering is not turned off, a midfrequency notch in the transducer's frequency response will result.

Bone conduction transducers used in devices coupled to the skin-covered skull are normally encased in a plastic housing. If the transducer is attached to the side of the housing pressed onto the skin, the housing itself does not affect the transducer characteristics and the output is primarily a function of the transducer design. However, if the transducer is attached to the side facing out from the skull, the housing is in the vibration transmission pathway and influences the transducer response. This housing response usually has a high-frequency resonance that increases the output from the transducer at the resonance frequency, but above this resonance frequency, the output of the transducer and housing falls off, limiting the bandwidth of the device. This is the case for the Radioear B71 diagnostic clinical bone conduction transducer, for which the housing resonance is usually between 3 and 4 kHz (it depends on the skin properties), limiting its performance at higher frequencies. The recently developed Radioear B81 transducer has less harmonic distortion and can be used at higher output levels, especially at lower frequencies. However, it still suffers from a housing resonance that limits the high-frequency performance and this is one reason why bone conduction thresholds typically are not measured above 4 kHz in the clinic.

The frequency response of the variable-reluctance transducer is to a large extent determined by its mass and spring properties. The resonance frequency is inversely proportional to the square root of the transducer mass and compliance and the lower this resonance frequency the better the low-frequency response of the transducer. To improve the low-frequency output of the transducer, the mass of the transducer can be increased or the stiffness decreased. However, there is an incentive to make hearing aids as small and light as possible to satisfy users' preferences. As a result, most bone conduction hearing aids have poor performance at low frequencies.

One of the drawbacks with the classic design of the variable-reluctance transducer is the amount of distortion produced, which limits its usability at higher output levels. The distortion is caused by the fact that the output force is related to the square of the magnetic flux. In a novel design using two static magnetic fluxes and air gaps, the distortion components generated at the two air gaps are opposite and cancel (Fig. 3.13a; Håkansson 2003). When compared with the Radioear B71 transducer, at 250 Hz and a level of 40 dB HL, the total harmonic distortion was reduced by more than 20 dB using this design (Håkansson 2003).

3.3.5.3 Piezoelectric Transducers

Another technology used for bone conduction transducers relies on the piezoelectric effect of certain materials, for example, quartz; an electrical field is generated when the material is forced to alter its dimensions. Correspondingly, applying a voltage to the material causes a change in its geometry. When an alternating current is applied to the material, the mechanical deformation is proportional to the current, and the transducer generates a mechanical vibration with high force but limited displacement. Using multiple crystals and stacking them onto each other (Fig. 3.13b), the total displacement is increased, and this is a common technique for increasing the efficiency of a piezoelectric transducer.

For the purpose of bone conduction excitation, one end of the piezo stack is connected to the skull (with or without skin). If the other end is free, the force applied to the skull is low because the reaction force is low, depending only on the small motion and mass of the piezo stack. To increase the applied force, a mass is usually attached to the free end of the piezo stack. The applied force is then basically the reaction force from the motion of the mass. As low frequencies require large displacements to generate the excitation force, a piezo stack transducer has poor low-frequency performance. However, it is very efficient in generating motion at high frequencies.

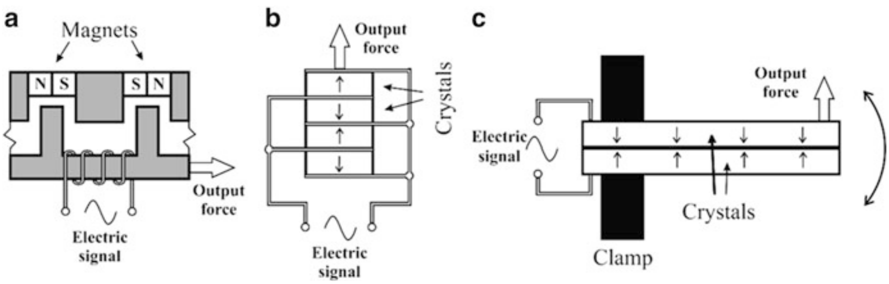

Fig. 3.13 (**a**) Drawing of the principle of the balanced electromagnetic separation transducer (BEST). The total air gap for the magnetic flux is constant, which makes the output more linear. (**b**) A schematic of a stacked piezoelectric transducer. The crystals are placed with opposite polarization (*arrows*) onto each other, resulting in a greater motion. (**c**) The piezoelectric bimorph uses two crystals attached with opposite polarization (series). With one end clamped, the beam is forced to bend, causing a motion at the other end

Another technique for increasing displacement with the piezo material is to attach two piezo beams and excite them in opposition; this is termed a piezoelectric bimorph (Fig. 3.13c). When excited in opposition, one beam elongates and the other shortens, causing a bending motion. To be effective, one end needs to be clamped and the other end forces the displacement. Based on a piezoelectric bending design, Adamson et al. (2010) proposed a bone conduction transducer that deformed the bone at the site of excitation to generate the vibration. Although such a design can be effective in generating bone conduction vibrations, it is not known if there is any long-term detrimental effect due to bone deformation.

3.3.6 Bone Conduction Coupling Issues

The operation of a bone conduction hearing aid depends greatly on its coupling to the skull. Historically, the bone conduction transducer was pressed onto the skin-covered head, usually at the mastoid, using a headband. With the advent of the titanium implant, percutaneous solutions became available in the 1980s and 1990s, while recently active subcutaneous bone conduction hearing aids and dental and ultrasonic devices for hearing by bone conduction have been introduced (Margolis and Popelka 2014).

3.3.6.1 Transcutaneous Devices

In a transcutaneous design, the transducer can be encased and the housing itself vibrates and is attached to the skin-covered bone (Fig. 3.14a). Alternatively, the transducer is attached to a flat surface that interfaces with the skin and transmits the vibrations to the skull (Fig. 3.14b). Regardless of which is used, a static force is required to press the bone conduction hearing aid or transducer against the skull. A common way to achieve the static force is to use a headband that presses the transducer or vibrating surface onto the head. The headband also holds the transducer in position. As stated previously, the greater the static force, the better the sound transmission. However, there are data suggesting that the static force has only a limited influence (Toll et al. 2011) and a large static force can cause discomfort. The static force is usually limited to about 2 newtons when a headband is used. To transmit bone conduction vibrations from the transducer to the skull, the static force needs to be greater than the dynamic force. A dynamic force of 2 newtons corresponds to 59 dB HL at 250 Hz, 84 dB HL at 1 kHz, and 95 dB HL at 2 kHz when the coupling surface area is 175 mm^2 and stimulation is at the mastoid.

The headband is usually a spring steel band that goes over the top of the head or an elastic band that goes around the head. The transducer is generally placed at the mastoid process, but it is sometimes placed on the forehead in young children. An alternative is to use glasses with a bone conduction transducer at the spring-loaded temple. Appropriate clients should need eye glasses as well as bone conduction hearing aids. That method is hardly used today, but other devices, such as Google Glass®, use bone conduction transducers at the temples to transmit bone conduction signals.

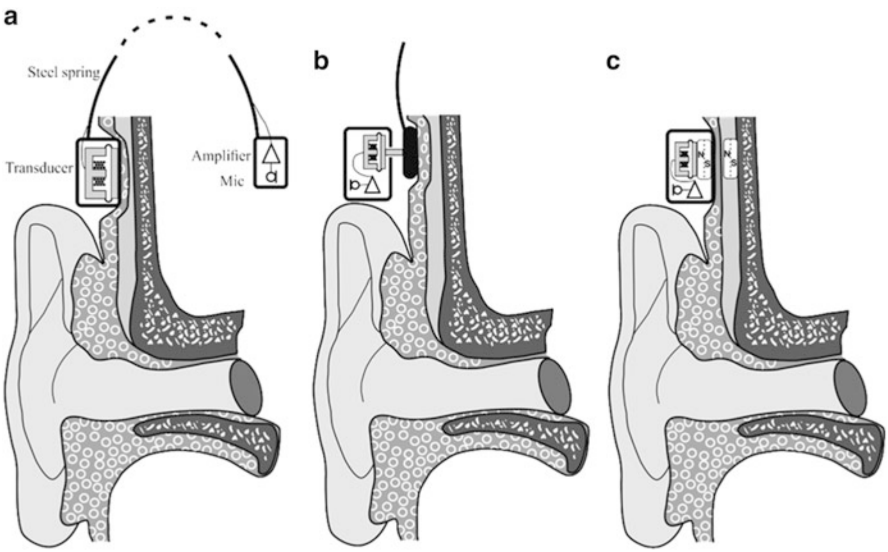

Fig. 3.14 (**a**) Schematic of a bone conduction hearing aid where the transducer presses onto the skin-covered mastoid with the static force from a steel spring. The microphone and amplifier are on the opposite side of the head, attached to the steel spring. The output from the amplifier is electrically wired to the transducer. (**b**) Principle of a single-house bone conduction hearing aid attached to a headband. The transducer is attached to a plastic plate that is pressed onto the mastoid using a soft band or a steel spring on the head. The microphone, signal-processing unit, amplifier, and transducer are encased in a single housing. (**c**) The bone conduction hearing aid is attached by magnets that provide the static force required to keep it in place. One of the magnets is surgically attached to the skull bone, while the other is attached to the transducer

Recently, magnets were introduced to provide the static force. The method requires that one part is implanted in the skull under the skin and the other part is placed onto the skin, integrated with the bone conduction transducer or as a separate plate (Fig. 3.14). Either the implanted part, the part on the skin, or both contain the magnet. This technology omits the headband, avoids the complications of a percutaneous post, and is more convenient for the user. A drawback with both magnetic systems and headbands is that they compress the skin and subcutaneous tissues and, over a long period of time, this can lead to necrosis of the tissue.

The attenuation produced by the skin and subcutaneous tissue for typical bone conduction hearing aids is estimated to average about 10 dB. This is consistent with estimates of the difference in effective stimulus between bone conduction applied at the skin surface and directly to the bone; the difference increases from almost zero at 0.25 kHz to approximately 20 dB at 8 kHz (Stenfelt and Håkansson 1999). This analysis is based on a constant voltage as the input to the transducer. If the applied force to the skull is used as the measure, the effect of the skin is mostly limited to 5 dB.

3.3.6.2 Percutaneous Devices

The attenuation produced by the skin is avoided when the stimulation is applied directly to the bone. This is achieved with a percutaneous titanium implant rigidly anchored in the parietal bone some 55 mm behind the ear canal opening (Snik et al. 2005). The implant is rigidly attached to the skull bone using a titanium screw that osseointegrates with the skull bone over time. A titanium fixture is attached to this implant, enabling direct vibration excitation of the skull bone through the covering skin (Fig. 3.15). In this case, the bone conduction transducer is attached to the fixture via a coupling, providing good vibration transmission up to 10 kHz.

Compared with transcutaneous devices using a headband, comfort is increased. Also, the direct coupling results in a wider bandwidth as well as higher maximum output levels. Percutaneous devices can therefore provide better speech perception, owing to the better high-frequency response and the fitting ranges are greater owing to the higher maximum output levels.

The transcutaneous hearing systems are primarily used for patients where no other option for hearing rehabilitation is possible. With the percutaneous bone conduction devices, patient satisfaction has increased and, when used within the limits of the

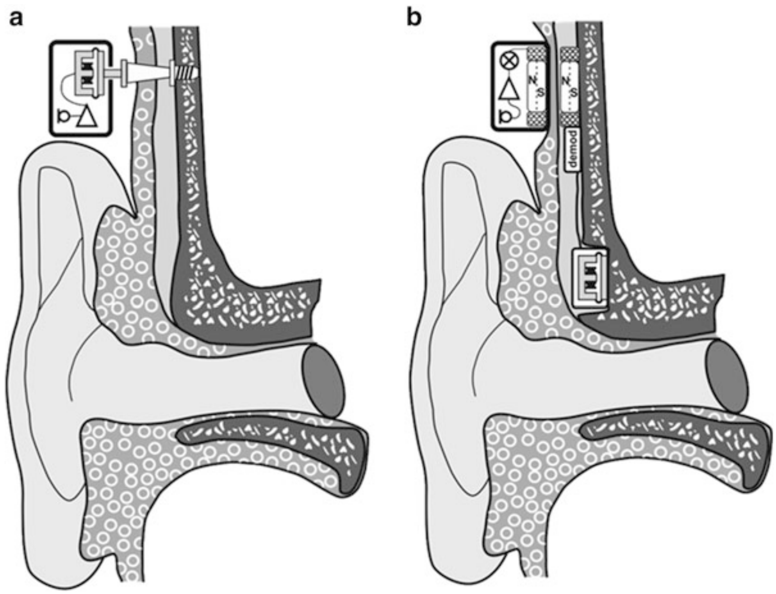

Fig. 3.15 (**a**) Illustration of a percutaneous hearing aid. The implant is surgically screwed into the cranial bone and a pedestal couples the external bone conduction hearing aid with the implanted screw. (**b**) Schematic of an implanted bone conduction hearing aid where the external unit is attached by magnets to the skull. The external unit consists of microphone, amplifier, signal processor, a radio frequency modulator, an induction coil, and a magnet. The internal unit consists of a magnet, induction coil, demodulator, and transducer

devices, the percutaneous systems are considered beneficial and are a well-accepted rehabilitation (Snik et al. 2005). Also, because the bone conduction pathway circumvents the outer and middle ear, the degree of conductive loss is not important. This is true for all bone conduction devices and the sensorineural component of the hearing loss determines the limit for usage of the different devices. At present, there are two manufacturers of percutaneous bone conduction hearing aid systems (Cochlear with the BAHA system and Oticon Medical with the Ponto system). Both have an ear-level power device with approximately the same maximum output characteristics: 40–45 dB HL at 0.25 kHz-increasing with frequency to approximately 80 dB HL at 2 kHz and decreasing to 75 dB HL at 4 kHz. Consequently, to have at least 30 dB dynamic range, the devices are appropriate only for maximum sensorineural hearing loss up to 45–50 dB HL at 1 kHz and above. The less powerful devices have about 6 dB less maximum output.

With an ear-level bone conduction hearing aid, the microphone, amplifiers, and transducer are in a single housing. Because the output from the transducer is a vibration and not an acoustic signal as in a conventional hearing aid, there is no direct acoustic feedback. However, there are several feedback pathways that can result in feedback problems. The vibrations from the transducer are coupled to the housing and can feed back to the microphone mechanically or acoustically from airborne sound radiated from the vibrating hearing aid housing. The third major route for feedback is from the skull vibrations that couple to the air, creating airborne sound that reaches the hearing aid microphone. A fourth possible feedback pathway is from magnetic signals from the transducer that can couple back to the microphone and input electronics, but the most problematic feedback pathways involve vibrations or radiated airborne sound.

3.3.6.3 Subcutaneous Active Devices

Drawbacks with skin-penetrating fixtures include daily care of the area around the fixture and infections in the skin surrounding the implant that may ultimately cause a loss of the implant. These issues have led to the invention of subcutaneous active bone conduction hearing aids. They use the benefit of direct connection to the bone of the skull while keeping the skin intact. The design of these devices is similar to that of cochlear implants or active middle ear implants, with the difference that the output is in the form of skull-bone vibration instead of electrical excitation of the auditory nerve or vibration of the middle ear ossicles. The key components of the subcutaneous active bone conduction hearing aid are an audio processor placed behind the ear, a wireless link that transmits the signal and energy to the internal component, a demodulator, and the bone conduction transducer (Fig. 3.15).

The audio processor is held in place on the skin using a magnet that aligns the coils used for the wireless transcutaneous transmission of the excitation signal. It is the electromagnetic signal that is transmitted through the skin, not the vibration signal. The loss from transmitting the electromagnetic signal wirelessly through the skin instead of directly coupling the signal amounts to about 10 dB. However, the placement of the bone conduction transducer closer to the cochlea improves the sensitivity of these devices by 5–15 dB. It has been shown that a position as close as possible to the cochlea is beneficial from a sound transmission point of view (Stenfelt and Goode 2005a; Eeg-Olofsson et al. 2008), but for practical reasons, the placement is in the mastoid portion of the temporal bone behind the external ear. That position results in a bone conduction hearing aid system with effective output similar to that of the common percutaneous systems (Håkansson et al. 2010; Huber et al. 2013).

3.3.6.4 Dental Devices

Dental applications for bone conduction hearing aids were proposed some time ago (Sabatino and Stromsta 1969; Dahlin et al. 1973), despite the fact that the oral cavity is an unfavorable environment for the transducer and electronics. One recent system uses the molars as the site for bone conduction excitation (Popelka et al. 2010b). The microphone and processor are placed externally at the ear and the signal is wirelessly transmitted to a receiver and transducer in a sealed housing clamped on either side of an upper back molar. As stated previously, the teeth provide a sensitive position for application of bone conduction stimulation because the upper jaw provides direct access to the skull bone (Stenfelt and Håkansson 1999). Applying the bone conduction signal via the teeth avoids artificial skin penetration, as the teeth themselves are rigidly anchored in the skull, and the system can be used without surgery.

The current system using dental application is intended primarily for persons with conductive hearing loss or with unilateral deafness. For unilateral hearing loss, the microphone and processing unit are placed at the deaf ear, the processed sound is transmitted wirelessly to the oral device coupled to the lateral surfaces of two molars, usually on the contralateral side of the head, near the healthy cochlea that is excited through bone conduction. Unilateral hearing loss can also be treated with the percutaneous devices described in Sect. 3.3.6.2, but then both the microphone and transducer are placed on the deaf side and the sound is transmitted to the healthy cochlea across the skull (Stenfelt 2005), resulting in a loss of sensitivity of up to 10 dB at the higher frequencies due to transcranial attenuation of bone-conducted sound (Stenfelt 2012). This transcranial attenuation is avoided with the dental device.

One problem with measuring the output of the dental and the subcutaneous active devices is that there is no artificial mechanical load for testing of the systems. The subcutaneous systems can probably use the same skull simulator measurement systems as used with the percutaneous devices (Håkansson and Carlsson 1989) because the mechanical loads are similar (Stenfelt and Goode 2005b). However, the dental application is different and no mechanical load for objective measures has been presented.

3.3.6.5 MRI Compatibility

Magnetic resonance imaging (MRI) is a widely used tool for medical imaging. The technology requires that the person under investigation is situated in a strong magnetic field (often 3 T or more). Magnetic materials experience forces when introduced in the magnetic field and this could potentially be dangerous to a person with an implanted pedestal or device constructed partly of magnetic materials. In addition, heating caused by induced currents is a potential problem. Other problems with magnetic materials in the MRI scanner are image artifacts when the structure of interest is close to the magnetic material or demagnetization of the magnets in the implanted device produced by the strong magnetic field.

Percutaneous systems with a titanium screw are relatively safe for use with MRI, once the external processor is removed. Subcutaneous active devices are more problematic. The forces on these implants can be hazardous and demagnetization may occur. Therefore, before an MRI exam, these implants may need to be surgically removed.

3.3.6.6 Children Versus Adults

The reference data used to assess bone conduction hearing, both hearing thresholds and mechanical impedance loads, are applicable to adults. Therefore, when bone conduction hearing aids are used in children, the effective output may be different from that for an adult. The data on bone conduction hearing in children and infants are limited, but sensitivity at lower frequencies seems to be higher for children than for adults (Cone-Wesson and Ramirez 1997; Hulecki and Small 2011). This indicates that, for children, a bone conduction hearing aid should be set to a lower gain for frequencies below 1 kHz to compensate for the increased sensitivity.

The reasons for the increased sensitivity in children are not clear, but the head size is smaller and the mass is less than that of an adult. That means that the bone conduction transducer may create larger displacements for a given vibration force and thereby greater bone conduction stimulation. Another anatomical difference is that in infants all skull sutures have not fused. The skull impedance is lower for a nonfused skull than for a fused skull, with the result that, for a given applied force, the displacement at low frequencies is greater in an infant than in an adult.

In small children and infants, only transcutaneous applications of bone conduction hearing aids are used. One of the reasons is that the skull bone is thin and it is difficult and unwise to implant a screw in that bone. Another reason is that in some children in whom a bone conduction hearing aid is indicated, reconstructive surgery may be an option and the percutaneous implant could make that surgery more difficult or impossible.

3.4 Summary

Within the last half century, there has been a reduction in microphone size of three orders of magnitude, while microphone noise has remained low enough to allow near-normal aided thresholds. The vibration sensitivity of microphones has also dropped by three orders of magnitude, and the vibration output of dual receivers has dropped by an order of magnitude. The result is that the available gain is no longer limited by the transducers but by the quality of the earmold seal. Similarly, the available bandwidth is not limited by the transducers but by the hearing aid amplifiers; 16-kHz bandwidth has been routine in some hearing aid designs for 25 years.

Improvements are ongoing. For example, MEMS microphones with low battery drain and excellent noise performance and a practical 1-mm-thick receiver became available in about 2012. At present, progress in transducers appears to be ahead of that in digital signal processing. For example, most digital hearing aid circuits have a bandwidth less than 10 kHz, and many such circuits overload at the modest levels encountered at live unamplified classical and jazz concerts (cf. Whitmer, Wright-Whyte, Holman, and Akeroyd, Chap. 10).

The progress is less impressive for transducers and couplings used for bone conduction. The major limitations for those devices are the maximum output level and the bandwidth. Even so, progress has been made with new subcutaneous active devices and with devices attached to places other than the temporal bone, such as the teeth. Devices designed to operate with intact skin may overcome the problems with skin penetration that have dominated the bone conduction hearing aid market the last 25 years.

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Conflict of interest Mead Killion has no conflict of interest.
Stefan Stenfelt declares that he has no conflict of interest.
Daniel Warren is an employee of Knowles Corporation, a manufacturer of hearing aid transducers.
Van Halteren is an employee of Sonion, which makes transducers.

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Chapter 4
Hearing Aid Signal Processing

Stefan Launer, Justin A. Zakis, and Brian C.J. Moore

Abstract This chapter reviews the general types of signal processing that are used in modern digital hearing aids. The focus is on concepts underlying the processing rather than on details of the implementation. The signal processing can be classified into three broad classes: (1) Processing to apply frequency- and level-dependent amplification to restore audibility and provide acceptable loudness, based on the hearing profile of the individual (usually the audiogram but sometimes taking into account the results of loudness scaling) and the preferences of the individual. Frequency lowering can be considered as an additional method for restoring the audibility of high-frequency sounds. (2) Sound cleaning, for example, partial removal of stationary noises or impulse sounds and reduction of acoustic feedback. Noise reduction may be achieved using both single-microphone and multiple-microphone algorithms, but only the latter have been shown to improve intelligibility. (3) Environment classification for automatically controlling the settings of a hearing aid in different listening situations. It is concluded that modern hearing aids can be effective in restoring audibility and providing acceptable loudness and listening comfort, but they are still of limited effectiveness in improving the intelligibility of speech in noisy situations.

Keywords Beamforming • Binaural beamforming • Compression speed • Directional microphone • Environment classification • Frequency compression • Frequency lowering • Multichannel compression • Noise canceler • Noise reduction • Pinna simulation • Reverberation canceler • Signal processing • Spectral change

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enhancement • Spectral enhancement • Time–frequency analysis • Transient reduction • Wind noise detection • Wind noise reduction

4.1 Introduction

The world has truly become digital: the size of microelectronics has shrunk and the computational capacity of digital microelectronics has doubled nearly every 2 years rather precisely following what is called "Moore's law of microelectronics." This has had a significant impact on hearing aid technology. After a long history of analog hearing aids, a group at Washington University in St. Louis, MO, conceived of and patented the concept of an all-digital hearing aid (Engebretson et al. 1985). This patent contained several key claims including a bidirectional interface with an external computer, self-calibration, self-adjustment, wide bandwidth, digital programmability, a fitting algorithm based on audibility, internal storage of digital programs, and fully digital multichannel amplitude compression and output limiting. This group created several prototype hearing aids using custom digital signal-processing chips with low power and very large scale integrated (VLSI) chip technology and used these hearing aids for research on hearing-impaired people. The first commercial fully digital hearing aids became available early in 1992. Now, virtually all commercial hearing aids are fully digital and their digital signal-processing capability has significantly increased. Figure 4.1 shows signal-processing flow charts for first-generation and current-generation signal processors. It is remarkable how dramatically the density of signal-processing building blocks as well as the interaction between building blocks has grown. Modern hearing aids have turned into intelligent systems offering a range of specific algorithms or algorithmic settings for addressing the specific listening and communication needs of users in different acoustic environments.

Many hearing aids today include both application-specific integrated circuits (ASICs) and reprogrammable digital signal-processing (DSP) cores and microcontroller cores that make it possible to flexibly reuse the same microelectronics for various different signal-processing algorithms or algorithmic settings. These platforms can offer a portfolio of signal-processing strategies depending on listening needs in specific environments. From the authors' perspective, the major performance improvement that has taken place over the past two decades stems from two lines of innovation (in addition to improvements resulting from wireless technologies, as discussed by Mecklenburger and Groth, Chap. 5). First, the focus of attention has broadened; although improving speech intelligibility in complex listening environments is still a major driver of innovation, it is no longer the only driver. Other aspects, such as sound quality improvement, reducing the occurrence of acoustic artifacts, and making the hearing aids sound natural, have become important too. Hearing aids have to function well in everyday life situations and not only for improving speech intelligibility in quiet or noisy environments. Sound scenes

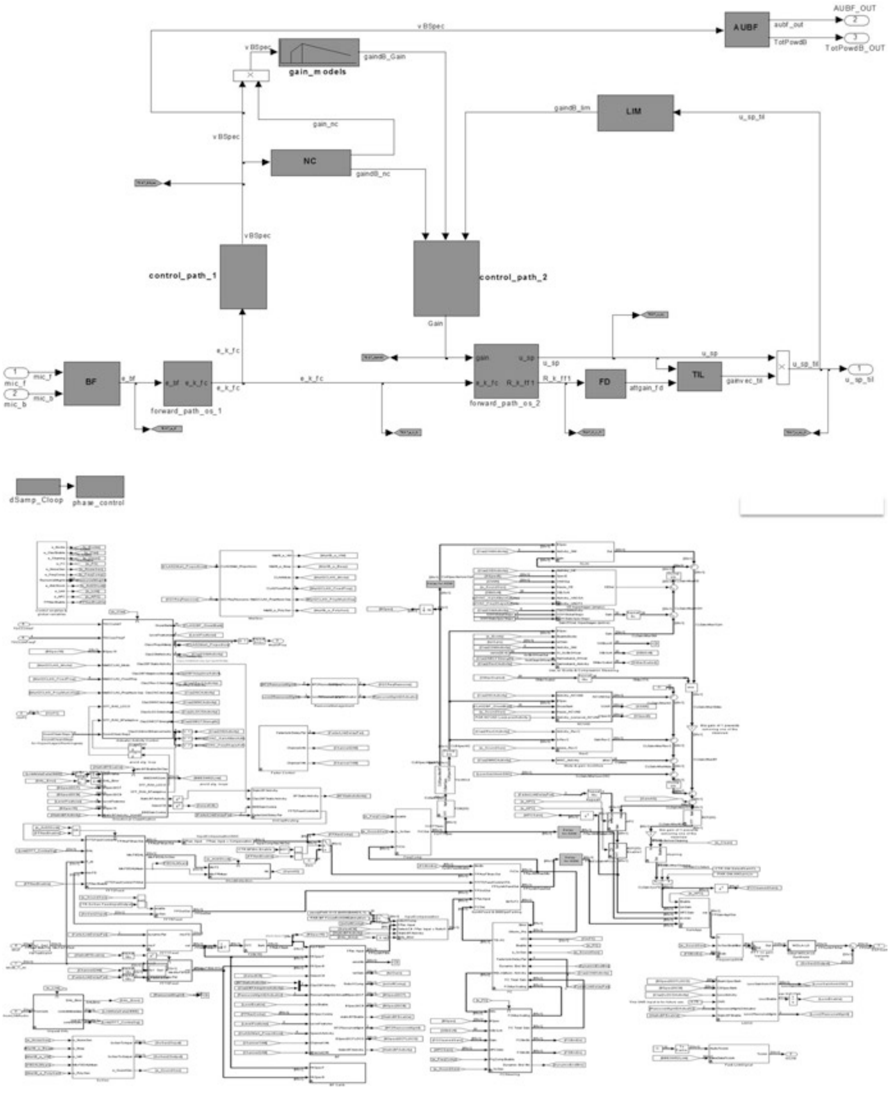

Fig. 4.1 Flow charts of the signal processing for first-generation (*top*) and current-generation (*bottom*) digital hearing aids. The text is not intended to be readable; the point of the figure is to indicate the jump in complexity

constantly change and evolve over time. Hearing aids should provide good sound quality throughout very dynamically changing listening environments. Modern DSP platforms provide the computational power to implement new functionality such as frequency lowering, impulse noise cancelers, and adaptive directional microphones, addressing a broadened range of listening needs in a variety of acoustic environments.

The second area of improvement is in systems integration. This is far better understood than it was and is taken into consideration when designing, implementing, and choosing parameters for signal-processing algorithms. Many of the adaptive algorithms that are commonly implemented in hearing aids can influence and sometimes counteract each other; for example, noise-canceling algorithms and algorithms for control of acoustic feedback tend to reduce gain and work in opposition to amplification schemes. System designers also take into account the mechanical design aspects of hearing aids. These can have an important impact on the performance of some algorithms. For example, the suspension and positioning of transducers affect the performance of acoustic feedback control systems (see Sect. 4.6.3) and the positioning of microphones in a behind-the ear (BTE) or in-the-ear (ITE) device affects the performance of directional microphones. Additionally, most of the algorithms used in modern hearing aids can be adjusted to improve their efficacy in different listening conditions. For example, an acoustic feedback canceler might be set very differently for optimal communication in quiet surroundings and for enjoying music.

Finally, a crucial issue is how to assess the performance and patient benefit of signal-processing algorithms in real-life environments. Assessments should include objective and subjective measures of patient benefit on dimensions such as speech intelligibility, listening effort, and sound quality as well as technical measures; see Munro and Mueller, Chap. 9 and Whitmer, Wright-Whyte, Holman, and Akeroyd, Chap. 10.

This chapter includes a short overview of the various types of signal-processing algorithms incorporated in modern hearing aids, followed by sections describing and discussing specific algorithms in more detail. The focus of this chapter is on current aspects and concepts rather than on technical details. For more detailed information, readers are referred to the books by Dillon (2012) and Kates (2008) and to the review papers by Hamacher et al. (2005, 2006) and Edwards (2007).

4.2 General Overview of Signal-Processing Algorithms

Figure 4.2 gives an overview of the various types of signal-processing algorithms available in modern digital hearing aids. The signals are picked up mostly by one or two omnidirectional microphones and converted from the analog to digital (A/D) domain (AD conversion) with an input dynamic range of up to 100 dB and sampling frequencies between 16 and 32 kHz, providing a usable audio bandwidth of 8–16 kHz; see also Moore and Popelka, Chap. 1; Killion, Van Halteren, Stenfelt, and Warren, Chap. 3; and Popelka and Moore, Chap. 11. Such a large dynamic range is important to be able to process the large ranges of sound levels encountered in everyday life and to provide good sound quality, especially for hearing-impaired people with milder hearing loss, who may be sensitive to low-level noise produced by AD converters and to harmonic and intermodulation distortion produced by output clipping. However, not all hearing aids achieve such a wide dynamic range; see Moore and Popelka, Chap. 1 and Zakis, Chap. 8.

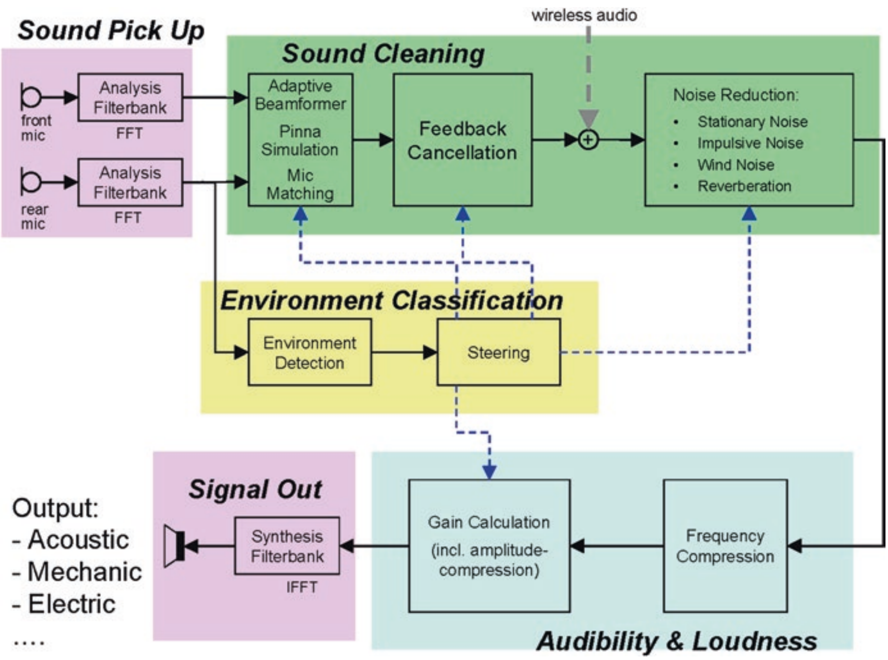

Fig. 4.2 Generic signal-processing scheme of a modern digital hearing aid

After sound pickup and AD conversion, the discrete time domain signal is analyzed spectrally using either a time domain filtering process or a transformation into the frequency domain. Both approaches involve a trade-off in time–frequency resolution. The signal delay should not be too long and rapid analysis entails coarse spectral resolution. However, spectral resolution should be sufficient for the subsequent signal-processing algorithms. Especially when using a hearing aid with an open fitting (see Chap. 1), a delay longer than 10–12 ms causes the sound to become "echoic" or "hollow" because of the interaction of the direct sound through the vent and the delayed processed and amplified sound (Stone and Moore 1999; Stone et al. 2008). Most current hearing aids have a total delay of 3–8 ms, which avoids deleterious effects of the delays, including mismatch between the audio signal and the facial and lip movements of the speaker but limits the spectral resolution of the time–frequency analysis. In most devices, spectral resolution depends on frequency, with the channel bandwidth being smallest at low frequencies (typically 80–150 Hz) and typically increasing roughly in proportion with the center frequency, similar to the way that auditory filter bandwidths vary with center frequency (Glasberg and Moore 1990).

Subsequent algorithms can be broadly classified according to their functionality:

1. Processing to apply frequency- and level-dependent amplification to restore audibility and provide acceptable loudness based on the specific individual hearing profile (usually the audiogram, but sometimes taking into account the results

of loudness scaling) and the preferences of the individual hearing-impaired person. Frequency lowering can be considered as an additional method for restoring audibility of high-frequency sounds.
2. Sound cleaning, for example, partial removal of stationary or nonstationary interfering sounds and reduction of acoustic feedback. One of the major problems of hearing-impaired listeners is the perception of a target signal in the presence of interfering sounds. One approach to improving speech intelligibility in such adverse or challenging listening conditions is to identify and reduce interfering sources. This can be achieved by applying either single-microphone or multiple-microphone noise reduction algorithms (other methods, such as the use of remote microphones, are described in Chap. 5).
3. Environment classification for automatically controlling the settings of a hearing aid.

Hearing aids from different manufacturers differ in the specific algorithmic solutions and range of signal-processing options they include. However, the broad classes of algorithms are the same. In each class, several different solutions exist. No single algorithm can provide optimal performance in the vast range of daily listening environments. Algorithms are usually designed on the basis of specific assumptions about the listening situation and signals to be processed. Environmental classifiers continuously monitor the listening situation and select appropriate algorithms and algorithm settings.

4.3 Time–Frequency Analysis

Spectral analysis can be performed in the time or frequency domain and "on-line" (i.e., in the signal path) or "off-line" (i.e., in a processing side branch). Alternative approaches involve different combinations of computational load, frequency resolution, time resolution, artifacts, and delay. These factors are traded off against each other to achieve a satisfactory combination within the practical limits of computational load and delay. The delay from spectral analysis must be considered in the context of delays from other digital processing that may exist. Analog and signal conversion circuits can further delay the signal by about 1–3 ms in total (Ryan and Tewari 2009; Zakis et al. 2012). The trade-offs of some alternative approaches to spectral analysis are discussed in Sects. 4.3.1–4.3.3. A good review of the various approaches can be found in Löllmann and Vary (2008).

4.3.1 Time Domain Analysis

Spectral analysis can be performed in the time domain with a bank of infinite impulse response (IIR) filters. An example is shown in Fig. 4.3 (top) for one microphone and is duplicated for the second microphone (not shown). Each filter creates

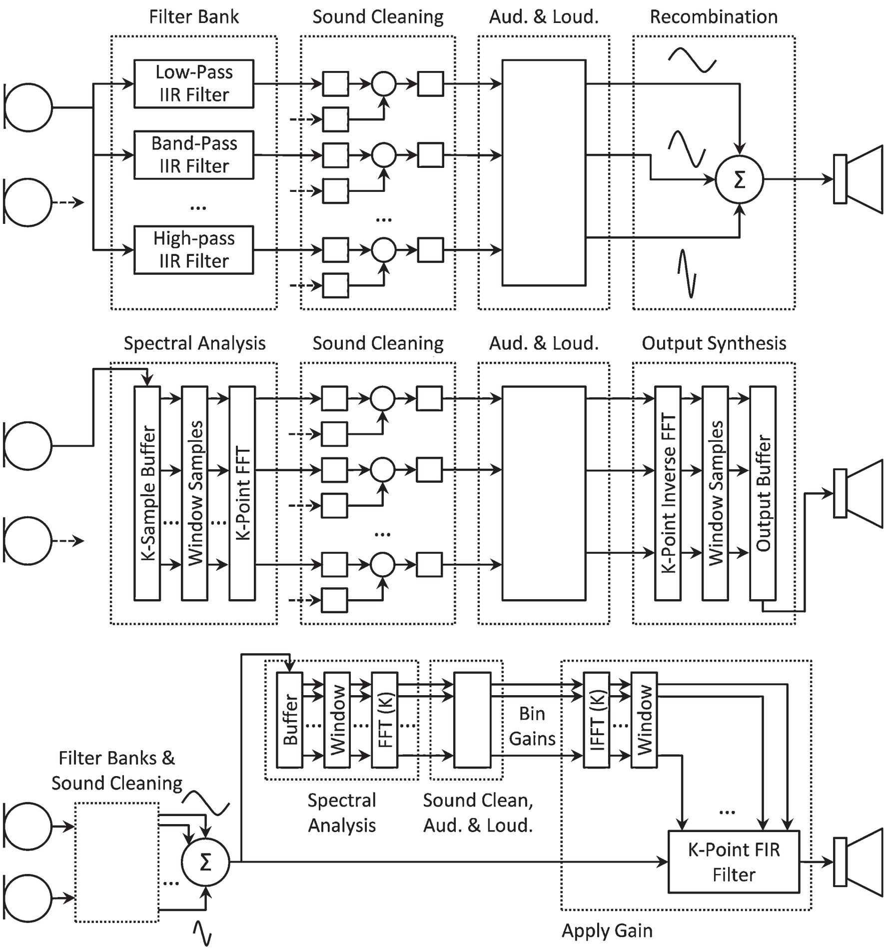

Fig. 4.3 Block diagrams detailing the spectral analysis and recombination performed in examples of time domain (*top*), frequency domain (*middle*), and hybrid (*bottom*) systems

one frequency channel. The number of channels varies widely across hearing aids, and the channel widths typically increase with frequency in a similar way as for the human auditory system (Glasberg and Moore 1990). The channel signals are processed by on-line, time domain versions of subsequent algorithms and then summed to form the output signal.

IIR filters achieve the desired filter slopes with fewer computations than finite impulse response (FIR) filters. However, careful filter bank design is needed to avoid unwanted effects (Kates 2005). For example, the phase responses of the IIR filters can lead to unintended peaks and valleys in the magnitude–frequency response when the outputs of the different filters are combined (after processing) to form the output signal. Furthermore, the frequencies at which the peaks and valleys occur can vary with the gain–frequency response and this may lead to audible artifacts.

For IIR filters, the delay generally increases with increasing slope and the delay also varies with frequency within the passband, being greatest at the filter's passband edge frequencies. In practical systems, the filter delay is of the order of 1 ms (Kates 2005). Steeper slopes also require more computations, which may limit the number of filters that can be implemented. It is more computationally efficient for filters (and subsequent algorithms) to process a block of samples at once rather than to process each sample as it arrives from the AD converter. However, each block of K samples must first be buffered prior to filtering, which delays the signal by an additional K samples.

4.3.2 Frequency Domain Analysis

Figure 4.3 (middle) shows a system with spectral analysis and subsequent algorithms in the frequency domain. Spectral analysis is shown for one microphone and is duplicated for the second microphone (not shown). The most recent samples are stored in an input buffer of length K. This block of samples is periodically multiplied by a windowing function that tapers off at its edges and is then converted to the frequency domain via a K-point fast Fourier transform (FFT). The windowing function reduces discontinuities at the ends of the block that adversely affect spectral analysis. The shape and duration of this window determine some properties of the filtering (Harris 1978). The FFT result consists of $(K/2)+1$ linearly spaced frequency bins (channels), which contain data that can be used to estimate the input level and phase in each bin. Subsequent algorithms can combine bins to form wider analysis channels (e.g., when it is desired that the channel bandwidth should increase with increasing center frequency) and modify the FFT result to clean and amplify the signal. An inverse FFT converts the modified FFT result to a block of K-modified samples. This block is added to samples from older blocks in the output buffer with a temporal overlap determined by the period between FFTs (Kates 2005).

FFTs efficiently divide the signal into a large number of frequency bins. As an example, a $K=128$-point FFT divides the spectrum into 65 frequency bins, which are centered at multiples of 125 Hz when the sampling rate is 16 kHz (Kates 2005). This is traded off against a time resolution of K samples (8 ms) and a delay of K samples (8 ms) due to buffering and windowing. If the FFT size is halved, time resolution and delay are improved by a factor of 2, but frequency resolution worsens by a factor of 2 (i.e., there are half as many bins that are twice as wide). Modifying the FFT result by a set of gains is equivalent to processing a block of L samples by an M-coefficient filter in the time domain, which results in $L+M-1$ processed samples (Kates 2005). When the number of processed samples is greater than the FFT size K, the processed samples "wrap around" the inverse FFT output and cause temporal aliasing distortion (Kates 2005). Reducing the number of samples in the input buffer (and zero padding, i.e., adding zeros on either side of the actual sample values) so that $L<K$, and/or smoothing the gain across frequency (which reduces M) can make this imperceptible (Kates 2005). Distortion can also occur with insufficient temporal

overlap between successive blocks. Greater overlap requires more frequent calculation of FFTs and inverse FFTs and hence increases computational load. FFTs and inverse FFTs require more multiply-and-add operations than time domain filters, which increases round-off error and hence quantization noise (Kates 2005). However, this can be minimized with careful design or the use of DSP circuits with larger word lengths (processing using a greater number of bits).

4.3.3 Hybrid Analysis Systems

Figure 4.3 (bottom) shows an example of on-line processing in the time domain combined with off-line processing in the frequency domain. The microphone signals are processed by IIR filter banks and by the two-microphone sound-cleaning algorithms (e.g., multiple-channel directionality; see Sect. 4.5 for details). The cleaned signal is summed across channels to form a single wideband signal prior to the off-line processing branch. This branch consists of an FFT filter bank and the sound cleaning and amplification algorithms that calculate a set of frequency-dependent gains. In this system, an inverse FFT converts these gains to a corresponding set of FIR filter coefficients. Thus, the on-line FIR filter applies the frequency-dependent gains of the off-line sound cleaning and amplification algorithms to the wideband signal.

Such systems combine the lower quantization noise and distortion of time domain filters with the higher frequency resolution of FFT systems (Kates 2005). As the output signal is not synthesized from an inverse FFT, it is not prone to distortion produced by the inverse FFT or by the nonperfect summation of successive blocks. Therefore, there is less need to smooth the gain across frequency or have large temporal overlap between successive blocks. Thus, frequency domain processing can be run less often, which frees up computational load for the on-line FIR filter.

FIR filters designed as described in the preceding text have a linear phase response, and the delay is $K/2$ samples for all frequencies. Therefore, if $K=128$, the FIR filter delay is 64 samples (4 ms with a sampling rate of 16 kHz). A signal is fully “seen” by the FFT and FIR filters about K samples after its onset, so the FIR filter’s gain is well synchronized with the signal it processes. The IIR filters and any on-line buffering of filter inputs (not shown) add to the FIR filter delay.

In an alternative approach, additional off-line processing can be used to design a minimum-phase FIR filter with a delay of about two samples irrespective of the FFT size (Dickson and Steele 2010). Thus, frequency resolution can be increased without increasing the FIR filter delay. The delay varies slightly with frequency (Zakis et al. 2012) and increases with steeper gain–frequency response slopes in a similar way to what occurs in analog systems (Kates 2005). The FIR filter gain now lags the signal by about $(K/2) - 2$ samples, which contributes to overshoot with fast-acting compression (see Sect. 4.4.1) but has little effect with slow-acting compression.

Another hybrid system uses digital frequency warping (Kates 2005; Kates and Arehart 2005). This involves modifications to the FIR filter structure and FFT input and results in nonlinearly spaced FFT bins that increase in width with increasing frequency. This avoids the need to group bins to form wider high-frequency channels and allows the use of a smaller FFT (better time resolution) and FIR filter, which compensates for the extra computational load of frequency warping. The delay is similar to that of a nonwarped system and progressively reduces with increasing frequency.

Time–frequency analysis schemes form the basis for applying the various adaptive algorithms that are discussed in this chapter. Owing to the various constraints discussed in Sects. 4.3.1 and 4.3.2, the time–frequency analysis used in hearing aids is rather limited compared to that performed in the human auditory system or in applications that do not require low-delay real-time audio processing. This in turn limits the performance of subsequent algorithms.

Time–frequency analysis schemes also provide the basis for the clinician to individually set some of the major processing parameters of the hearing aid to the individual needs of a hearing-impaired person (see Chap. 9). The number of analysis channels used for signal-processing purposes often exceeds the number of channels that a clinician can realistically be expected to modify during the fitting procedure. Currently, there is uncertainty about the optimal number of channels for signal processing and the optimal number that should be made available to the clinician for adjustment to suit the needs of the individual patient.

4.4 Signal Processing for Restoring Audibility

4.4.1 Multichannel Compression

Providing audibility of sound while avoiding uncomfortable loudness is a crucial role of digital hearing aids. Amplification schemes have become more sophisticated over the past 15 years (Kates 2008; Dillon 2012). Besides providing appropriate audibility and loudness in everyday listening conditions covering a wide range of sound levels, the amplification schemes have to be designed so as to provide good sound quality with as few artifacts and as little distortion as possible. Because the amount and pattern of hearing loss varies markedly across individuals, amplification schemes and their parameters need to be adjusted to match the frequency-specific needs of the individual. Most people with sensorineural hearing loss experience loudness recruitment; once the level of a sound exceeds the elevated absolute threshold, the loudness grows more rapidly than normal with increasing sound level (Fowler 1936; Moore 2007). On average, the greater the hearing loss, the greater the rate of growth of loudness (Miskolczy-Fodor 1960). However, individual variability can be considerable. Typically, hearing aids process sounds in 10–20 frequency channels, the width of the channels increasing with increasing center frequency. In each of these channels, a level-dependent gain is applied. Almost any gain-versus-level function can be applied using digital processing.

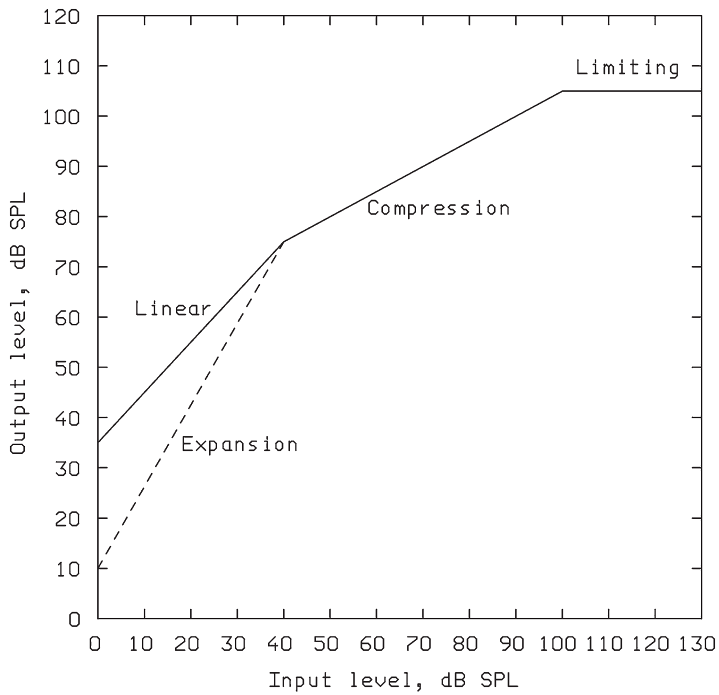

Fig. 4.4 Schematic input–output function for one channel of a multichannel hearing aid. Three level regions are illustrated. For input levels from the compression threshold (CT, 40 dB SPL in this example) up to 100 dB SPL, compression is applied. For input levels above 100 dB SPL, limiting is applied. For input levels below the CT, either linear amplification (*solid line*) or expansion (*dashed line*) may be applied

To compensate for loudness recruitment, the gain should decrease progressively with increasing input level, meaning that the input–output function is compressive. When compression is applied, an increase in input level of X dB gives rise to an increase in output level of Y dB, where $Y<X$. The ratio X/Y is called the compression ratio.

Owing to technical limitations of transducers and microelectronic circuits, especially at low and high input levels, compression cannot be applied over the entire level range. Basically, one can distinguish three level ranges with different gain characteristics, as illustrated in Fig. 4.4. For very low input levels, up to about 40 dB SPL, the gain may be kept constant (linear processing), but more often the gain increases with increasing level, that is, expansion rather than compression is applied. This is done to prevent internal noises such as microphone noise or the intrinsic noise floor of the microelectronic circuits from being audible. For input sound levels from about 40 to 100 dB SPL, the applied gain decreases with increasing level so as to compensate for loudness recruitment. For very high input levels the compression ratio is made near infinite. This is called “limiting” and it leads to a maximum output sound level that can be set separately for each channel, usually based on the highest comfortable level for the patient. Details of the level-gain characteristic may

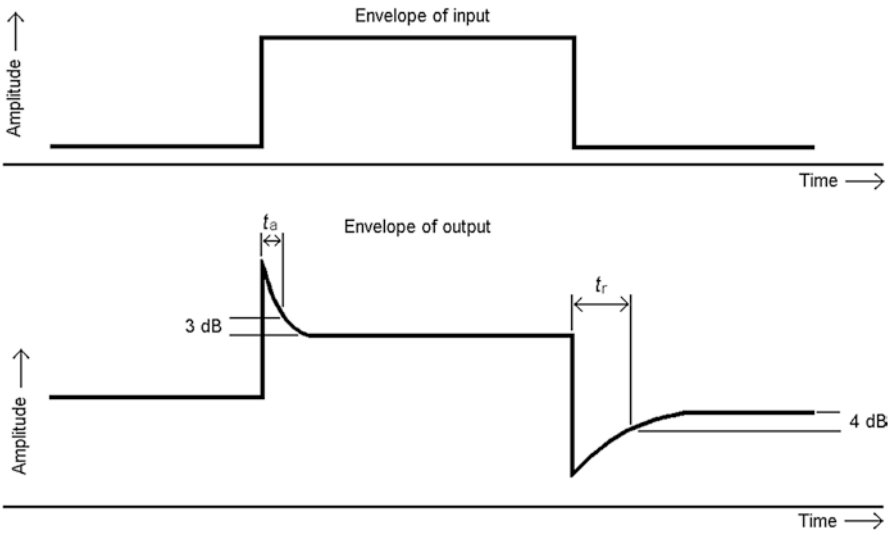

Fig. 4.5 Illustration of the effect of abrupt changes in level at the input to a hearing aid. The envelope of the input is shown at the *top* and the envelope of the output is shown at the *bottom*. The output shows an "overshoot" when the level suddenly increases and an "undershoot" when the level decreases

differ strongly between hearing aids, but the concept of having three level regimes, "expansive" at low levels, "compressive" at medium levels, and "limiting" at high levels, is common to most modern hearing aids. The characteristics of the level-gain curves have to be adjusted to suit the hearing loss and the listening preferences of the individual patient; see Chap. 9.

The levels of everyday sounds can change rapidly over time. Amplitude compression systems vary in how quickly they react to changes in sound level. Typically, the speed of response is measured by using as an input a sound whose level changes abruptly between two values, normally 55 dB SPL and 90 dB SPL. When the sound level abruptly increases, the gain decreases, but this takes time to occur; if the change in gain were made instantaneous, this would produce audible distortion in the waveform. Hence the output of the compressor shows an initial "overshoot," followed by a decline to a steady value, as illustrated in Fig. 4.5. The time taken for the output to get within 3 dB of its steady value is called the attack time and is labeled t_a in Fig. 4.5 (ANSI 2003). When the sound level abruptly decreases, the gain increases, but again this takes time to occur. Hence the output of the compressor shows an initial dip, followed by an increase to a steady value. The time taken for the output to increase to within 4 dB of its steady value is called the recovery time or release time and is labeled t_r in Fig. 4.5 (ANSI 2003).

Compression systems in hearing aids can be divided into two broad classes. The first is intended to adjust the gain automatically for different listening situations. Such systems relieve the user of the need to adjust the volume control, which may be important for older people with limited manual dexterity. The gain is changed slowly with changes in input sound level; this is achieved by making the recovery time, or both the recovery time and the attack time, relatively long (usually t_r is between 0.5 and 20 s). The compression ratio in such systems can be high (if the design philosophy is to present all sounds at a comfortable level) or more moderate (if the design philosophy is to give some impression of the overall level of sounds in the environment). One specific implementation of a slow-acting system is called "adaptive dynamic range optimization" (ADRO). Note that ADRO does not base the gain on the input level but rather adjusts the gain based on a comparison of the output level to output targets. This is described in Chap. 8.

The second class of compression system is intended to make the hearing-impaired person's perception of loudness more like that of a normally hearing listener. Because loudness recruitment behaves like fast-acting multichannel expansion (Moore et al. 1996), restoration of loudness perception to "normal" requires fast-acting multichannel compression. Systems with this goal have relatively short attack and recovery times (t_a is 0.5–20 ms and t_r is 5–200 ms). They are often referred to as "fast-acting compressors" or "syllabic compressors" because the gain changes over times comparable to the durations of individual syllables in speech. Fast-acting compressors usually have lower compression ratios than slow-acting systems. High compression ratios (above about 3) are avoided, as these have been shown to have deleterious effects on speech intelligibility (Verschuure et al. 1996; Souza 2002).

Both slow-acting and fast-acting forms of compression have advantages and disadvantages. The advantages of slow-acting compression are:

1. If desired, signals can be delivered at a comfortable loudness, regardless of the input level, by use of a high compression ratio.
2. The temporal envelopes of signals are only minimally distorted. This may be important for maintaining speech intelligibility (Stone and Moore 2003, 2004).
3. Short-term changes in the spectral patterns of sounds, which convey information in speech, are not distorted because the pattern of gains across frequency changes only slowly with time.
4. Short-term level changes are preserved, so cues for sound localization based on interaural level differences are not markedly disrupted (Moore et al. 2016); see Akeroyd and Whitmer, Chap. 7.

The disadvantages of slow-acting compression are:

1. Loudness perception is not restored to "normal."
2. It may not deal effectively with situations in which two voices alternate with markedly different levels, for example, when one talker is nearby while another is farther away.
3. When there is a sudden drop in sound level, for example, when moving from a noisy bar to a quiet room, the gain takes some time to increase. Hence the aid may appear to become "dead" for a while, and some soft speech may be missed.
4. When trying to listen to a target voice in the presence of background voices, a normally hearing person can extract information about the target during the temporal dips in the background (Duquesnoy 1983); see Souza, Chap. 6.

This process is called "listening in the dips." The information in the dips may be at a relatively low level. Slow-acting AGC is of limited benefit in this situation because the gain does not increase significantly during brief dips in the input signal.

The advantages of fast-acting compression are:

1. It can make loudness perception closer to "normal." However, normal loudness perception is not quite achieved. When a person has loudness recruitment, an amplitude-modulated sound appears to fluctuate more than normal (Moore et al. 1996). This is true for modulation rates up to at least 32 Hz. Even at the short end of the range of time constants used in hearing aids, fast-acting compression does not reduce the depth of amplitude modulation for rates above about 10 Hz (Stone and Moore 1992, 2003, 2004; Moore et al. 2001). Thus, dynamic aspects of loudness perception are not fully restored to normal.
2. If many subbands are used, fast-acting compression can compensate for frequency-dependent changes in the degree of loudness recruitment more effectively than slow-acting compression. While slow-acting compression can apply gain that is appropriate for the average level of the signal in each subband, fast-acting compression can also compensate for the short-term changes in signal level.
3. Fast-acting compression can restore the audibility of weak sounds rapidly following intense sounds. This provides the potential for listening in the dips (Moore et al. 1999).
4. When two voices alternate with markedly different levels, fast compression can make the softer voice audible without the more intense voice being unpleasantly loud.

The disadvantages of fast-acting compression are:

1. It can introduce spurious changes in the shape of the temporal envelope of sounds (e.g., overshoot and undershoot effects) (Stone and Moore 2007), although such effects can be reduced by delaying the audio signal by a small amount relative to the gain-control signal (Robinson and Huntington 1973; Stone et al. 1999).
2. It can introduce spurious changes in amplitude of sounds gliding in frequency, such as formants in speech, as those sounds traverse the boundary between two channels. This happens mainly for systems in which the compression channels are formed using sharp, nonoverlapping filters. The effect does not occur for systems in which the filters used to form the channels have frequency responses that overlap and have rounded tops and sloping edges (Lindemann and Worrall 2000).
3. It reduces intensity contrasts and the modulation depth of signals, which may have an adverse effect on speech intelligibility (Plomp 1988).
4. In a hearing aid with fast-acting compression in many channels, the spectrum is flattened. This compounds difficulties produced by the reduced frequency selectivity that is associated with hearing loss (Moore 2007).
5. When the input signal to the compressor is a mixture of different voices fast-acting compression introduces "cross-modulation" between the voices, because the time-varying gain of the compressor is applied to the mixture (Stone and

Moore 2004, 2007). This may decrease the ability to perceptually segregate the voices.

6. When moderate levels of background sound are present (e.g., noise from ventilation and air conditioning systems), fast compression makes such sounds audible, and this can be annoying (Laurence et al. 1983). When the number of subbands is small, steady background noises may appear to be modulated by "foreground" sounds such as speech. This can also be annoying. However, this effect is reduced when the number of channels is increased.
7. Cues for sound localization based on interaural level differences may be disrupted by the independent action of the compression at the two ears (Van den Bogaert et al. 2006; Wiggins and Seeber 2013); see Chap. 7. This effect can be avoided by synchronization of the compressor action across the two ears (Wiggins and Seeber 2013).

Modern hearing aids vary markedly in compression speed from one manufacturer to another and even within manufacturers. In some hearing aids, the compression speed can be adjusted. Some hearing aids incorporate dual time-constant systems that combine slow- and fast-acting compression (Moore and Glasberg 1988; Stone et al. 1999). There is evidence for individual differences in preference for compression speed (Gatehouse et al. 2006a, b; Lunner and Sundewall-Thoren 2007), but the factors underlying these preferences are poorly understood.

4.4.2 Recoding of High-Frequency Information: Frequency Lowering

It is often difficult to restore audibility at high frequencies for people with severe or profound hearing loss. The gains required can be very high and the use of high gains can lead to acoustic feedback and/or distortion and possible damage to residual hearing. In addition, the dynamic range of people with severe or profound hearing loss can be very small, making it difficult to avoid loudness discomfort when high gains are used. An alternative approach is frequency lowering, in which the high frequencies in the input are shifted to lower frequencies where the audiometric thresholds are not so high. This recoding of information makes it easier to restore audibility and to avoid loudness discomfort. Also, it may be perceptually advantageous to present signal information at frequencies where a person has a more moderate hearing loss rather than where the hearing loss is severe or profound, as auditory processing may be better in the region of moderate loss (Moore 2007).

Frequency-lowering techniques were intensively studied from 1960 to 1980 (Braida et al. 1979) and interest was rekindled through the work of Hugh McDermott and co-workers (Simpson et al. 2005a, b, 2006) on frequency compression; see below for details. For reviews giving details of the various approaches, see Robinson and Moore (2007), Simpson (2009), and Alexander (2013). Over the years many different forms of frequency-lowering techniques have been evaluated. These can be broadly classified in three groups, as illustrated in Fig. 4.6: (1) frequency transposi-

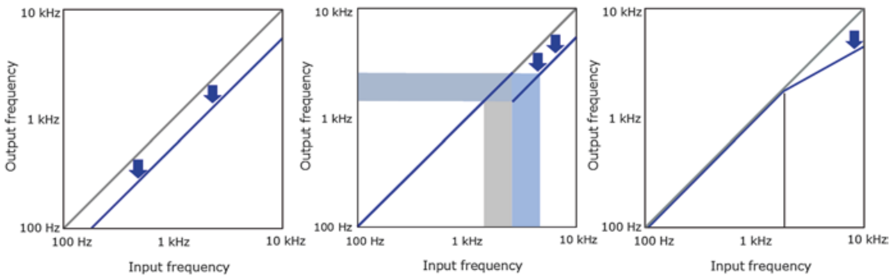

Fig. 4.6 Schematic illustration of different frequency-lowering schemes: full bandwidth transposition, (*left*) partial transposition, (*center*) and partial frequency compression (*right*)

tion; (2) frequency compression or frequency division; and (3) lowering of part of the spectral envelope, keeping the spectral components unaltered. The first two approaches shift signal components from higher frequencies to lower frequencies, either over the entire spectral range or only over part of the range. Frequency lowering over the entire spectral range was applied only in the very first approaches because it changes the fundamental frequencies of speech sounds and thus severely alters the quality and perceived identity of voices.

4.4.2.1 Frequency Transposition

In this approach (Robinson et al. 2007; Kuk et al. 2009), a block of higher frequency components (the source band) is shifted downward in frequency to a destination band. The two bands have the same width in hertz. The transposed energy is usually superimposed on the energy that is already present in the destination band. The source band may be fixed, or it may be selected to fall around the dominant peak in the high-frequency part of the signal spectrum for a specific time frame. The transposition may be "conditional" in that it occurs only when the signal in a specific time frame has a relatively high ratio of high-frequency to low-frequency energy, probably indicating the presence of a fricative in speech (Robinson et al. 2007).

4.4.2.2 Frequency Compression

With frequency compression, frequency components up to a "starting frequency" remain unchanged in frequency, and frequency components above that frequency are shifted downward by an amount that increases with increasing frequency. For example, a source band from 4 to 8 kHz may be compressed into a destination band from 4 to 6 kHz (Simpson et al. 2005a). In this approach, the frequency-lowered energy is not superimposed on energy from another frequency region, but the destination band is narrower than the source band. This class of algorithms modifies the spectral relationship of the signal components. Specifically, harmonic tone-like signals such as vowels become inharmonic after application of the frequency

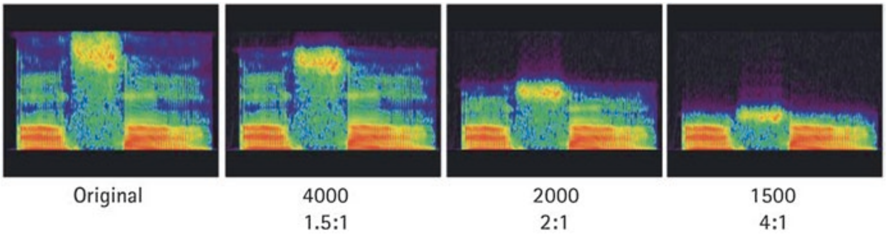

Fig. 4.7 Spectrograms of "asa" processed with different amounts of frequency compression. In the spectrograms, time is on the *x*-axis (range = 500 ms), frequency is on the *y*-axis (100–10,000 Hz, logarithmic scale), and the brightness of the color denotes intensity. The upper numbers below the *x*-axis indicate the cutoff frequency (Hz) and the lower numbers indicate the frequency compression ratio

compression. Usually, the starting frequency is chosen to be relatively high (above 1.5 kHz), as even normally hearing listeners are relatively insensitive to inharmonicity in the frequency region above 1.5 kHz (Darwin and Carlyon 1995). A specific form of frequency compression was used by Simpson et al. (2005a), defined by the following equation

$$F_{out} = F_{in}{}^{\alpha}, where\ \alpha < 1$$

where F_{in} is the input frequency, F_{out} is the output frequency, and α represents the amount of frequency compression applied. The effect of frequency lowering is illustrated in Fig. 4.7.

4.4.2.3 Lowering of the Spectral Envelope

In this approach, sometimes called "spectral envelope warping," the spectral envelope is modeled using "linear predictive coding" (Schroeder and Atal 1985). This is based on the assumption that the signal is speechlike and that the spectral envelope can be modeled as a series of resonances and antiresonances. The spectral envelope at low frequencies is not changed, but the envelope at high frequencies is progressively shifted downward. With this approach, the spectral "fine structure" is not altered and harmonic relationships are preserved. The shifted spectral information is superimposed on the original spectrum. The processing may be activated only when appropriate high-frequency spectral features are detected, which is a form of conditional transposition.

4.4.2.4 Clinical Evaluations of Frequency Lowering

Several studies of the efficacy of frequency compression algorithms have shown modest benefits at the group level but distinct benefits for some individual participants (Simpson et al. 2005a; Glista et al. 2009; Wolfe et al. 2010). Participants rarely perform worse with frequency compression. Typically, participants with more severe hearing loss seem to benefit from frequency compression, whereas for participants with milder hearing losses the results are ambiguous—some studies report benefit (Wolfe et al. 2015) while others show less benefit (Picou et al. 2015). Also, the benefits tend to be larger for children than for adults (Glista et al. 2009). Typically, a long acclimatization period is not needed to gain benefit in consonant perception when using frequency compression (Hopkins et al. 2014).

It is not clear whether frequency compression provides benefit for speech production and language development. Ching et al. (2013) assessed the development of speech and language at 3 years of age for two groups of children, one using conventional amplification and one using frequency compression. Receptive and expressive language scores were higher but receptive vocabulary and consonant articulation scores were lower for children who used frequency compression. There was increased substitution of affricates by fricatives for children using frequency compression. Ching et al. concluded that there is insufficient evidence to indicate a difference in language ability between children using frequency compression and those using conventional amplification. Bentler et al. (2014) also found no difference in speech and language development between children using conventional amplification and those using frequency compression.

Somewhat different results were obtained by Zhang et al. (2014). They assessed (1) perception of the high-frequency consonant /s/; (2) cortical responses; and (3) functional performance as judged by parents. The children, aged 2–7 years, were tested first without frequency compression and then after 6 weeks of use of frequency compression. With frequency compression, the children showed improved perception of the consonant /s/ and increased cortical responses to /s/. Ratings of functional performance were also improved. However, the experimental design of this study is problematic, as the results could have been affected by learning and maturation effects independent of the signal processing.

Most studies testing frequency-lowering algorithms showed benefit for some individual participants but no study provided a clear indication of which participants benefit from frequency lowering and which do not. Developing a candidacy profile for frequency lowering remains an open need.

4.5 Signal Processing to Improve Speech Intelligibility in Challenging Listening Conditions

One of the most challenging situations for hearing-impaired people is understanding target speech in background sounds; see Chap. 6. Usually the term "speech in noise" has been used to describe these types of conditions but this fails to reflect the huge variability of the speech signals and background sounds encountered in daily listening and communication situations. The target speech varies because of the characteristics of the talker, the vocal effort of the talker, the distance of the talker from the listener, the orientation of the talker relative to the listener, the amount of reverberation in the listening environment, and whether or not visual cues are available. The background sounds vary in spectrum, in level, in whether they are relatively steady or strongly fluctuating, in their distance from the listener, in how they are spatially distributed, and in the amount of reverberation. Owing to this variability, no single optimal noise reduction algorithm exists. However, a range of algorithms can be used to provide benefit in different listening conditions. This section outlines algorithms that are in use today.

4.5.1 Directional Microphones

The most successful approach is based on the use of directional microphones, which can help when the target speech is spatially separated from the interfering sounds. Hearing aids often have two omnidirectional microphones, one located toward the front of the aid and one toward the back. By delaying the signal from one microphone and adding the outputs of the two microphones, one can create a “beam pattern” that points in a specific, usually frontal, direction; see Chap. 3. In this way, target signals from the direction of the beam will be picked up well while sounds from the side or rear will be attenuated. By varying how the outputs of the two microphones are delayed and combined, different beam patterns can be generated (Widrow and Luo 2003; Elko and Meyer 2008). This is often called “beamforming.”

In a static beamformer, the shape of the beam pattern is constant, while in an adaptive beamformer, the shape of beam pattern dynamically adapts depending on the environment and especially the direction of the most prominent interfering sounds. When the direction of a prominent interfering sound changes, such a system will maintain good suppression of the interference. Another possibility is to implement the beamformer independently in different frequency bands rather than applying it to the broadband signal. This can improve the performance of the beamformer when the interfering sources differ in spectral content.

Directional microphones have been extensively tested in laboratory environments, mimicking a large number of different real-world environments and using a range of different tests including different spatial distribution of interfering sources, different reverberation times, and different types of beamformers; for reviews see Ricketts (2001) and Dillon (2012). The typical benefit observed in realistic listening conditions corresponds to a 2–5 dB improvement in the speech reception threshold (e.g., the signal-to-background ratio required for 50 % correct) in a speech-in-noise test.

Although directional microphones with beams pointing to the front do bring clear benefits, they also have some limitations. One is that the target speech may come from the side or even the back (e.g., in a car), in which case directional microphones would make it more difficult to hear the target speech. Another limitation is related to the placement of the microphones on the ear. Optimal beamforming can be achieved by ensuring that the microphones are well matched in amplitude and phase response as a function of frequency and that their ports are horizontally aligned and point to the front. Tilting the direction toward the ceiling, for example, by positioning a BTE aid too far behind the pinna, significantly reduces the directivity that can be achieved. For cosmetic reasons, BTE aids are often positioned well down behind the pinna so as to make them nearly invisible, but this comes at the expense of reduced directivity. Another factor affecting the performance of directional microphone systems is the acoustic coupling applied in a BTE device. For mild to moderate hearing losses, an open fitting is usually used (see Chap. 1) and this limits the effect of the beamformer at lower frequencies. Beamformers in BTEs with open-fitting domes still show significant benefits, but they are not as large as those obtained with closed molds (Magnusson et al. 2013).

4.5.2 Binaural Beamformers

An extension of the approach described in Sect. 4.5.1 is to combine the four microphones of two bilaterally worn hearing aids to form a four-microphone directional system. First the outputs of the two microphones on each side are processed to obtain a standard front-facing beam (this has a directional pattern described as cardioid; see Chap. 3). Then these independently processed directional signals are exchanged over a wireless link with the hearing aid on the other side; see Chap. 5. Utilizing a frequency-dependent weighting function, each hearing aid then linearly combines the ipsilateral and contralateral signals to create a binaural directivity pattern. The binaural beam width is controlled by the weighting function and is typically narrower than can be achieved with a monaural two-microphone beamformer. This static binaural beamformer can be extended to an adaptive binaural beamformer that adapts the binaural directivity to the current spatiotemporal distribution of interfering sounds. This is accomplished by adaptively combining the static binaural beamformer output with a directional signal calculated from the ipsilateral and contralateral microphone signals. Picou et al. (2014) have shown that such an approach can improve speech intelligibility and reduce listening effort compared to what is obtained with monaural directional microphones.

4.5.3 Binaural Better Ear

The use of conventional directional microphones is based on the assumption that the listening target is directly in front of the listener. However, in many communication situations, this is not the case. For example, when driving in a car, the target is usually on the side of the listener. A solution for listening environments where the target is not in front of the listener is to pick up the signal on the side of the "better" ear, that is, the ear ipsilateral to the target, and transmit it to the contralateral ear. In this way, both ears receive a reasonably "clean" representation of the target. Wu et al. (2013) have shown that such an approach can be beneficial for hearing-impaired listeners when listening in a simulated car environment.

4.5.4 Single-Microphone Noise Cancelers

Single-microphone noise canceling as applied in digital hearing aids relies on the assumption that speech signals have different temporal properties than noise signals. It is usually assumed that background interfering sounds do not have strong low-rate temporal fluctuations, whereas speech signals do. The temporal fluctuations in different frequency bands can be measured and used to estimate the signal-to-interference ratio in each band. Various methods have been proposed for reducing interference based on the estimated signal-to-interference ratios. For a review, see Loizou (2013). Current hearing aids mostly use computationally efficient algorithms based on either spectral subtraction or Wiener filtering.

The assumptions about the temporal properties of target speech and interfering signals described earlier in this section limit the listening situations in which one can expect benefit from single-microphone noise cancelers to those where the background sound is reasonably stable, for example, the noise from air conditioning or from a large crowd of talkers. Despite this, single-microphone noise reduction schemes are applied in almost all current digital hearing aids.

Evaluations of single-microphone noise cancelers have shown that they can improve sound quality and subjective ease of listening, but they do not usually improve speech intelligibility as measured using standard clinical tests (Ricketts and Hornsby 2005; Bentler et al. 2008). Improved ease of listening could be important as it could lead to hearing-impaired people being better able to follow a conversation over a longer time with less effort (Sarampalis et al. 2009).

4.5.5 Reverberation Canceler

In enclosed environments, there are many transmission paths between a target source and a listener due to reflections of the sound from walls, floors, ceilings, and objects in the room. Moderate to strong reverberation, such as occurs in large rooms, staircases, or long hallways, can have a severe impact on the ability of both normally hearing and hearing-impaired listeners to perceive speech (Helfer and Wilbur 1990; Beutelmann and Brand 2006). Furthermore, the performance of beamformers typically worsens with increasing distance between target and listener and increasing amount of reverberation (Ricketts and Hornsby 2003). Mathematically, the mixture of direct and reflected sounds at each ear represents a convolution between the original signal and an unknown room impulse response. “Dereverberation” is a computationally complex problem (Hunag et al. 2007) because, for each ear signal, it is necessary to estimate two highly dynamic signals, the source signal, typically speech, and the room impulse response, which varies as the source and listener move. Most current approaches to dereverberation are too complex to implement in hearing aids and involve time delays that would be unacceptable for hearing aids.

A much simpler approach was taken by Lebart et al. (2001). They simply attenuated parts of the signal that were estimated to be dominated by reverberation. The signal was filtered into frequency channels. In each channel, parts of the signal that were decaying in amplitude in an appropriate way (based on the estimated reverberation time in each subband) were treated as "reverberation tails" and were attenuated. This algorithm has not been shown to improve speech intelligibility as measured with clinical speech tests but has been shown to improve sound quality and ease of listening in reverberant environments (Fabry and Tchorz 2005).

4.5.6 Reducing Front–Back Confusions

Sounds reflected from the folds of the pinnae interfere with the sounds that directly enter the ear canal to create complex patterns of peaks and dips in the spectrum. These patterns provide cues for sound localization and especially for resolving front–back confusions; see Chap. 7. The pinnae also provide some shielding from high-frequency sounds coming from behind the listener. When BTE hearing aids are used, the microphones are located above and behind the pinnae, and this eliminates the cues normally provided by the pinnae, leading to front–back confusion (Keidser et al. 2006; Van den Bogaert et al. 2011) and reducing spatial naturalness. Such confusion may be important for speech intelligibility in spatially complex environments, as intelligibility in such environments is partly limited by "informational masking" (see Chap. 6) and informational masking can be reduced using perceived location (Freyman et al. 1999; Kopco et al. 2010).

In its overall effect, the pinna resembles a microphone with an omnidirectional characteristic at low frequencies (up to about 1 kHz) and a directional (front-facing) characteristic above 1 kHz. This can be simulated using the two microphones in a BTE aid and frequency-specific signal processing, thus partially restoring the functionality of the pinna. This is illustrated in Fig. 4.8. Reintroducing pinna cues in this way has been shown to reduce the number of instances of front–back confusion made by hearing-impaired subjects and to provide a more natural perception of the environment (Jensen et al. 2013; Kuk et al. 2013).

4.5.7 Limitations of Methods for Improving Intelligibility in Adverse Listening Conditions

As has been discussed in the previous sections, signal processing in hearing aids offers a number of solutions for improving speech intelligibility in various adverse listening conditions. However, limitations remain. Depending on the environment, the performance of noise reduction schemes can vary markedly. Furthermore, the benefits of noise reduction and directional microphones and/or beamforming found

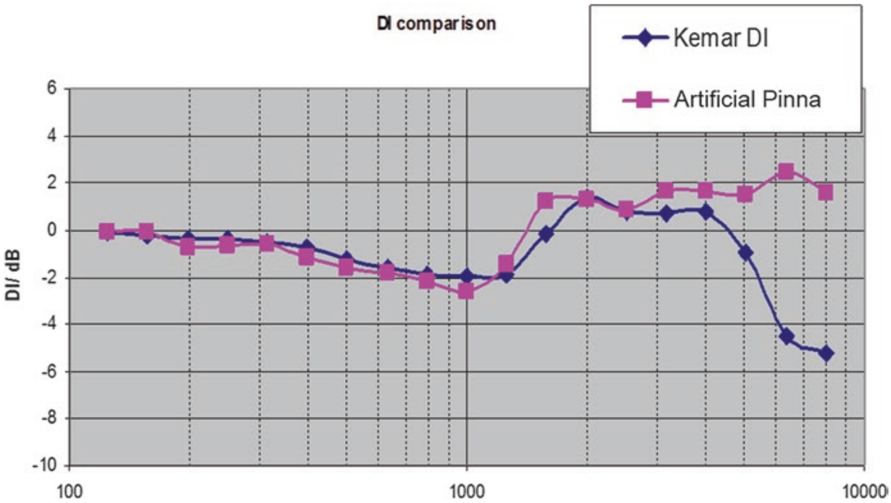

Fig. 4.8 Directivity index (DI) of the pinna on Kemar and for the pinna-restoring function in a BTE aid. The higher the DI, the greater the directivity

using questionnaires based on daily life situations (see Chap. 10) are often smaller than would be expected from experiments conducted in the laboratory. This may indicate that the assumptions used in the design of various algorithms are not always valid under realistic conditions.

4.6 Processing of Sounds to Improve Comfort and Sound Quality

Hearing aids are often worn for 14–18 h daily and the user hears all everyday sounds through them—not only speech sounds. The natural and authentic presentation of environmental sounds (including music; see Chap. 8) has become an important design goal in improving the performance of modern hearing aids, and this hopefully will contribute to better acceptance and wider usage of hearing aids by hearing-impaired people. Sections 4.6.1–4.6.3 review the types of signal processing that can be used to improve the comfort and naturalness of sounds.

4.6.1 Impulse Noise Canceler

Many sounds in everyday life have very impulsive characteristics. Examples include a door slamming, putting dishes on a table, a bat hitting a ball, a bell ringing, and the sound of a xylophone. Even when fast-acting wide dynamic range compression is

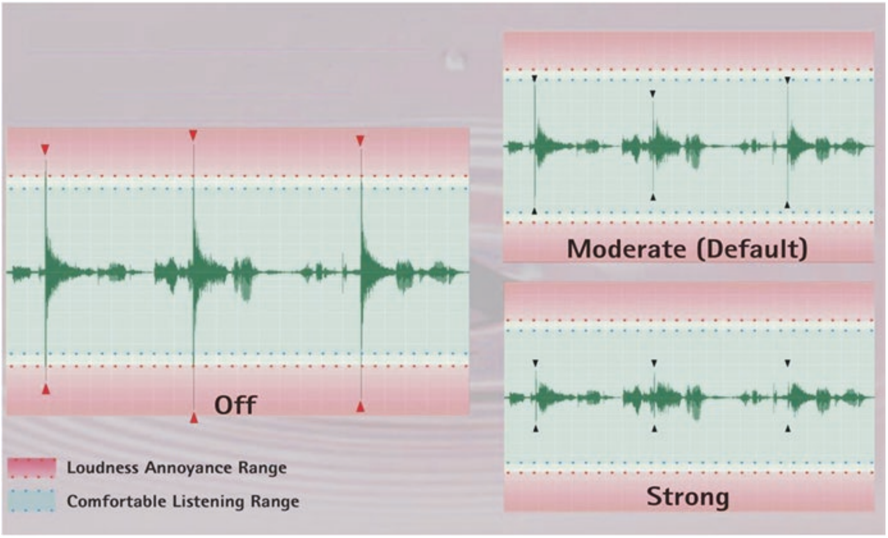

Fig. 4.9 Effect of an algorithm that reduces the level of intense transient sounds. For the original signal (*left*), strong annoying impulses are indicted by the *red triangles*. For the processed signal (*right*), these are reduced, as indicated by the *black triangles*

used, such impulsive sounds can be excessively amplified and cause annoyance or discomfort to a hearing aid user (Moore et al. 2011). To avoid this, many current hearing aids incorporate methods for detecting very fast increases in level and instantly applying a gain reduction for a short time. These algorithms are not intended to cancel the transient sound completely but rather to attenuate it so that the sound quality of the transient sound is authentic but the sound is not annoying. The operation of such an algorithm is illustrated in Fig. 4.9. Korhonen et al. (2013) used a paired-comparison task to compare the sound quality and annoyance of impulsive everyday sounds, such as a knife on a plate, a pen tap, and a car door, with an impulsive transient noise canceler on versus off. Experienced hearing aid users clearly preferred the algorithm on because the quality of the sounds was less annoying and more natural. Speech intelligibility was not adversely affected by the algorithm.

4.6.2 Wind Noise Reduction

Wind noise is caused by air pressure fluctuations resulting from turbulence. Turbulence can be intrinsic to wind outdoors (Morgan and Raspet 1992) and can develop as air flows around an object. Wind noise is a problem rarely noticed by people with normal hearing because the pinna and ear canal reduce the noise created by wind at the eardrum. However, when a hearing aid microphone is placed behind the pinna or close to the entrance to the ear canal, wind noise can be much more intrusive. Large, medium, and small objects such as the head, pinna, and microphone ports generate turbulence that causes low-, mid-, and high-frequency wind noise, respectively (Kates 2008). Wind noise increases in level and extends to higher frequencies with increasing wind speed (Zakis 2011). Wind noise levels vary with wind azimuth relative to the microphone port (Chung et al. 2009, 2010), across styles of hearing aid depending on whether the microphones are located behind the pinna or in the ear canal (Chung et al. 2010; Zakis 2011), and across models of the same style with different shell and microphone inlet designs (Chung et al. 2009; Zakis 2011). A wind speed of 12 m/s is sufficient for wind noise to saturate omnidirectional microphone circuits, causing a spread of the spectrum across the entire hearing aid bandwidth and leading to equivalent input levels of up to 116 dB SPL (Zakis 2011). For wind speeds below those producing saturation, wind noise levels are typically greater at the output of a directional than an omnidirectional microphone since turbulence is poorly correlated between microphone ports (Chung et al. 2009; Chung 2012a). The unique characteristics of wind noise can be exploited by detection and suppression algorithms.

4.6.2.1 Wind Noise Detection

Wind noise detection algorithms can operate in the time or frequency domain, can analyze the wideband signal or the signal in each channel of a multichannel system, and can process one or two microphone outputs (Kates 2008). Algorithms are broadly classed as either single or dual microphone. The choice of domain involves a trade-off between time and frequency resolution (see Sect. 4.3). Detection in individual channels allows better targeting of wind noise suppression processing across frequency but at the cost of greater computational load than for wideband detection. Multiple algorithms may be used together to improve detection reliability (Latzel 2013).

With a single microphone, detection is typically based on identification of the typical spectral characteristics of wind noise. Long-term average wind noise levels tend to be maximal at low frequencies and to decrease with increasing frequency at a fairly predictable rate (Kates 2008). However, wind noise spectra may not always fulfill these criteria (Chung 2012a) and the spectra may flatten after microphone saturation (Zakis 2011). Also, stationary background noise may obscure the wind noise spectrum. Finally, the use of long-term average levels limits the rapidity of detection.

With two microphones, detection is based on short-term comparisons of the microphone outputs. Acoustic waves from far-field sounds (e.g., speech) tend to be similar across microphones, while turbulence and hence wind noise tends to be independent across microphones. Therefore, wind noise can be detected when the microphone outputs differ substantially in level and/or phase (Elko 2007; Petersen et al. 2008). However, substantial differences in microphone outputs may occur due to the absence of wind, acoustic reflections, sound wavelengths that approach the microphone spacing, microphone port obstructions (hair, scarf, hat, etc.), and/or near-field sound sources (e.g., a telephone receiver held closer to one microphone than the other). For high wind speeds, differences between microphone outputs are reduced by microphone saturation. Such effects can reduce the distinction between wind noise and some non-wind sounds, which could lead to false positives and/or negatives in wind noise detection. Alternative approaches that compare waveform statistics across microphones over a block of samples can be more reliable, as they ignore the size of the level differences and are less sensitive to small phase differences (Zakis 2013; Zakis and Tan 2014).

4.6.2.2 Wind Noise Suppression

Several approaches can be taken separately or in combination to reduce wind noise independently in each hearing aid. One approach is to switch from a directional to an omnidirectional microphone, preferably only for wind-affected channels so the advantages of directionality are retained for other channels (Chung 2012a). Gain reductions based on the wind noise level and/or signal-to-wind noise ratio in each channel can increase comfort and potentially reduce the masking of speech by the wind noise (Latzel 2013). Other algorithms, such as stationary noise reduction (see Sect. 4.5.5), can be used to reduce wind noise, while multichannel compression can increase or reduce wind noise levels, depending on the magnitude of the wind noise (Chung 2012b).

Wireless links between hearing aids (see Chap. 5) have enabled the development of binaural wind noise reduction algorithms. One such algorithm transmits the microphone output from the less to the more wind-affected hearing aid when binaural wind noise asymmetry is detected (Latzel 2013; Latzel and Appleton 2013a). The more wind-affected aid uses the signal from the less wind-affected aid for frequencies below 1.5 kHz and its own signal for higher frequencies. This improves the signal-to-wind noise ratio at low frequencies and preserves sound localization cues at higher frequencies. As a result, speech understanding and listening effort are improved (Latzel and Appleton 2013b). Binaural algorithms could help address the traditionally low satisfaction with hearing aids in dealing with wind noise (Kochkin 2010a, b).

In summary, wind noise reduction is important both for reducing the annoyance produced by wind noise and for improving speech intelligibility and the audibility of other sounds (such as music) in windy conditions. Algorithms for reducing wind noise help the user of hearing aids to use them throughout the day, thus increasing overall acceptance.

4.6.3 Acoustic Feedback Management

Owing to the close proximity between the microphone(s) and the receiver in hearing aids and the high gain that may be applied by the hearing aids, acoustic instabilities or "feedback" can occur: the hearing aid output can leak back to the input via mechanical, electronic, and/or acoustic feedback paths. The focus of this section is acoustic feedback, which is the most problematic type for air conduction hearing aids. In most situations, acoustic feedback tends to occur for relatively high frequencies, from 1,500 to 8,000 Hz. Figure 4.10 (top) shows a simplified time domain signal path (bold arrows) consisting of a microphone, preamplifier (PA), analog-to-digital converter (ADC), DSP core, digital-to-analog converter (DAC), and receiver. Assuming for simplicity that these hardware components have unity gain (they do not in practice), the signal path gain is the processing gain $G(f)$ (f denotes frequency). The input sound $X(f)$ is amplified by gain $G(f)$ and becomes output $Y(f)$. Some of the output follows the acoustic feedback path with attenuation $H(f)$ and becomes feedback sound $F(f)$ at the microphone. This path is a combination of sound leakage paths that include the vent and gaps around the earmold. The fed-back output $F(f)$ is reamplified by gain $G(f)$, which completes the feedback loop and the cycle is repeated. At frequencies where the loop phase delay is an integer multiple of 360° and when gain $G(f)$ exceeds attenuation $H(f)$, the feedback is reinforced and increases in level every time it is reamplified. This results in a whistling or squealing sound. If $G(f)$ is equal to $H(f)$, whistling does not occur, but the sound quality is metallic and reverberant. The more $G(f)$ is below $H(f)$, the better is the sound quality. As the loop phase delay moves away from an integer multiple of 360°, increasing gain is required before these undesirable effects occur.

Prior to the advent of digital hearing aids, feedback reduction strategies included using a tighter (better fitting) mold, a smaller vent, and/or less high-frequency gain, which could also reduce physical comfort, the sound quality of the user's own voice, and speech intelligibility, respectively. The need for such strategies is reduced by digital feedback management algorithms, which can be categorized as either suppression or cancellation and can be used separately or in combination.

Feedback suppression algorithms reduce the gain at frequencies where feedback is detected. This is done in narrow bands to minimize reductions in speech intelligibility and sound quality. Gain reductions can be applied permanently for bands where feedback was problematic during fitting and/or adaptively for bands where feedback becomes problematic as the feedback path changes (Chung 2004). Feedback path changes can result from jaw movements that alter sound leakage around the earmold and/or reflections from nearby objects such as a hat, wall, or telephone handset. Reflections can increase the feedback level by more than 20 dB (Hellgren et al. 1999; Stinson and Daigle 2004). Feedback detection is typically based on searching for tonal inputs with the expected level and/or phase characteristics. Sounds such as music can have similar characteristics, and several strategies have been devised to reduce the chance of incorrect detection and/or suppression (Chung 2004).

Feedback cancelers (FBCs) subtract an estimate of the fed-back sound from the input to remove the actual feedback from the signal path. This increases the gain that can be used without whistling. However, estimation of the actual feedback signal is complicated by the fast gain changes that can be produced by amplitude compression systems and by noise reduction systems. Figure 4.10 (top) shows a simple FBC (below the simplified signal path). From the FBC's perspective, the DAC,

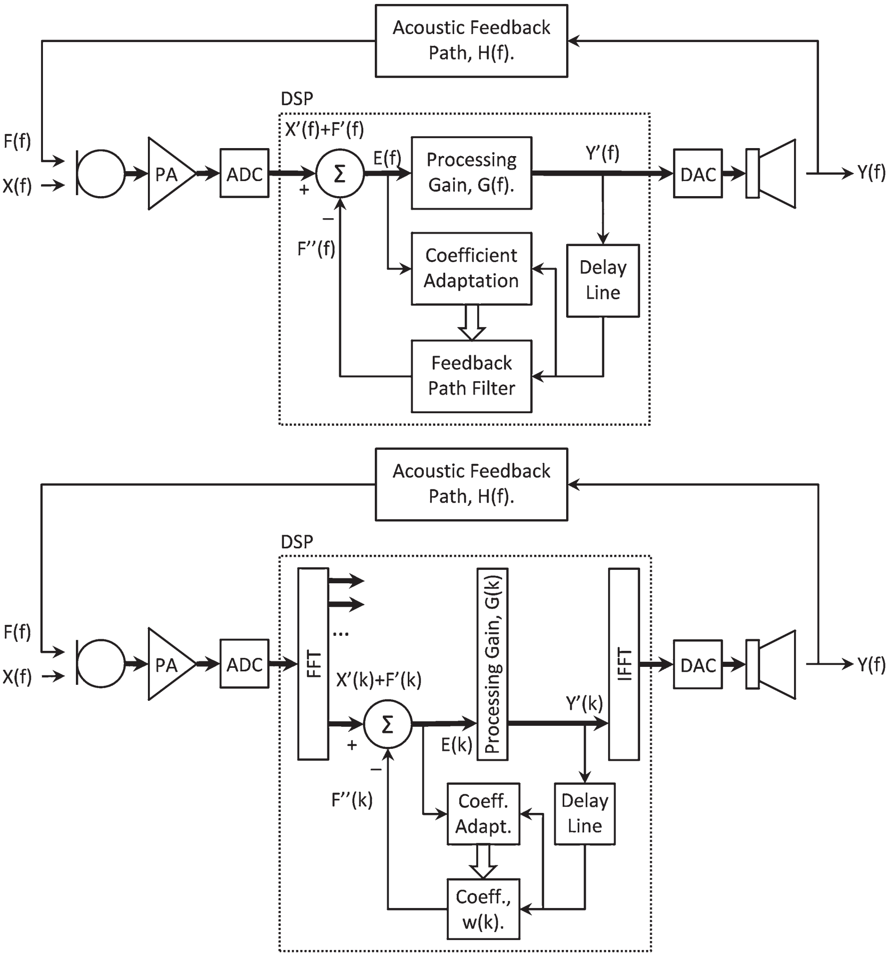

Fig. 4.10 Block diagrams of time domain (*top*) and frequency domain (*bottom*) schemes for acoustic feedback cancellation

receiver, microphone, preamplifier, and ADC are part of the feedback path. In practice, these circuits delay the audio signal and modify its magnitude- and phase-frequency responses. Thus, the FBC sees modified signals $X'(f)$, $F'(f)$, and $Y'(f)$. Output $Y'(f)$ is processed by a feedback path model that consists of a delay line and an FIR filter. The delay line simulates the feedback path delay, which is mainly due to the circuits. The filter is adapted to simulate the magnitude- and phase-frequency response of the feedback path; see later in this section for details of the adaptation. Filter output $F''(f)$ is an estimate of the actual feedback signal $F'(f)$. Feedback is more effectively removed from the input as the estimation error approaches zero.

Figure 4.10 (bottom) shows a frequency domain version of the above FBC algorithm for one FFT bin (the algorithm is duplicated for each bin). The FFT result represents the magnitude and phase of the input in bin number k and is modified by processing gain $G(k)$ to produce output $Y'(k)$. A single coefficient $w(k)$, which represents a gain and a phase shift, is adapted to model the feedback path over bin k's frequency range. Delayed output $Y'(k)$ is modified by coefficient $w(k)$ to produce feedback magnitude and phase estimate $F''(k)$, which is subtracted from the input to remove the actual feedback $F'(k)$.

In practice, up to 25 dB of additional gain can be achieved without whistling depending on the FBC algorithm and evaluation methods (Ricketts et al. 2008; Spriet et al. 2010). Perfect cancellation cannot be achieved for a number of reasons. (1) The adapted coefficients are stored with a finite number of bits, which introduces rounding errors that limit the accuracy of the feedback path model. (2) The room reverberation component of the feedback path cannot be adequately modeled with a practical number of coefficients, and reverberation limits model accuracy for the ear-level component (Kates 2001). (3) Nonlinearity in feedback path circuits (e.g., saturation) cannot be simulated with a linear model and reduces model accuracy for the linear part of the feedback path (Freed 2008). (4) The feedback path can vary rapidly, making it difficult to estimate. (5) The feedback is mixed with other sounds at the input, which increases the difficulty of identifying the feedback and hence modeling the feedback path.

A version of the least mean squares (LMS) algorithm (Widrow et al. 1976) is typically used to adapt the coefficients owing to its high computational efficiency. One trade-off is that although faster adaptation speeds increase model error, they also give better feedback control when the feedback path changes. The LMS algorithm is based on the assumption that signals that are correlated between the delayed output Y' and the postcancellation input E are feedback; the algorithm adapts the coefficients to minimize these signals. However, output Y' can be highly correlated with the desired input X owing to the low delay of the signal path processing. This is particularly likely for tonal inputs such as beeps or music. In this case, the filter is adapted to remove the desired input instead of the feedback. This problem is commonly known as entrainment and it causes an amplitude modulation artifact and less feedback control.

Various approaches have been used to avoid entrainment either separately or in combination (Guo et al. 2013). A slow adaptation speed reduces entrainment to brief tonal inputs but also slows adaptation to feedback path changes. Solutions for the latter include using a fast feedback suppression algorithm to control feedback during slow adaptation and switching to a fast adaptation speed when feedback is detected (Chung 2004). Another approach is to constrain the amount of adaptation around a good set of coefficients that were established during fitting (Kates 1999). This limits entrainment to tonal inputs but may also prevent full adaptation to large feedback path changes. Other approaches involve decorrelating the output from input X. This can be achieved by applying a constant (Joson et al. 1993; Guo et al. 2013) or time-varying (Freed and Soli 2006) frequency shift to the output. Larger shifts are more effective but are more likely to affect perceived pitch. Also, if an open fitting is used (see Chap. 1), interaction between the unshifted sound leaking to the eardrum and the shifted sound produced by the hearing aid may lead to beats and other undesirable effects. The LMS algorithm can be applied only at high frequencies. This targets adaptation to frequencies where feedback control is needed and prevents adaptation in response to correlated low-frequency sounds (Chi et al. 2003). Because correlation is stronger for frequencies close to spectral peaks, feedback can be further reduced with fixed filters that flatten the speech spectrum (Hellgren 2002) or adaptive filters that flatten any spectrum (Spriet et al. 2005) at the input to the LMS algorithm. Running a FBC algorithm separately in each frequency band limits the effect of correlation in one band on feedback path adaptation in other bands.

In summary, feedback suppression or cancellation algorithms increase the gain that can be used in hearing aids before acoustic feedback occurs. This makes it easier to achieve target gains, even for people with severe hearing loss, and it also increases the range of hearing losses that can be treated using open-fit hearing aids. The performance of feedback cancelers can be improved by measuring individual feedback path functions during the fitting process. However, feedback suppression systems are limited in their effectiveness and feedback cancellation system can produce undesirable side effects, so there is certainly room for improvement in FBCs. Furthermore, FBCs need to be carefully integrated into the entire signal-processing system and need to be connected and aligned with other adaptive algorithms such as those used for dynamic-range compression or noise cancellation. When designing an FBC system, the challenge is in finding the right trade-off between the aggressiveness of the FBC, the acoustic artifacts resulting from the FBC, and the increased gain that can be achieved via use of the FBC. Comparisons of the performance of different feedback cancellation systems need to take a holistic or systems perspective and not consider only the increase in gain that is achieved.

4.7 Environmental Classification and Control

As outlined in the introduction, modern hearing aids often offer a broad range of signal-processing strategies designed to provide good hearing performance in different listening conditions. Algorithms such as noise reduction systems or acoustic feedback management systems can be switched on and off or modified to change their behavior to deal with different listening situations. A classic example is when the hearing aid user moves from a quiet listening environment to a noisy restaurant. In the quiet condition, the microphone characteristic should be omnidirectional to allow detection of sounds from all directions, whereas in the noisy restaurant, a directional microphone characteristic would provide better speech intelligibility. Listening to music might require different settings from listening to speech, and the optimal settings might even depend on the type of music.

Many hearing aids allow multiple "programs" to be set up for different listening situations. These can be selected via a button on the hearing aid or via a remote control. However, some people find this tedious, and some people forget what the different programs are intended for. To avoid the need for manual switching, some automatic "environment control" algorithms have been introduced into modern hearing aids. These algorithms extract a variety of acoustic features and classify a listening environment by comparing the observed values of the features with a prestored map of values (Kates 1995; Nordqvist and Leijon 2004); see Fig. 4.11. The classifiers may employ a hidden Markov model, maximum likelihood, Bayesian estimators, or a "neural network."

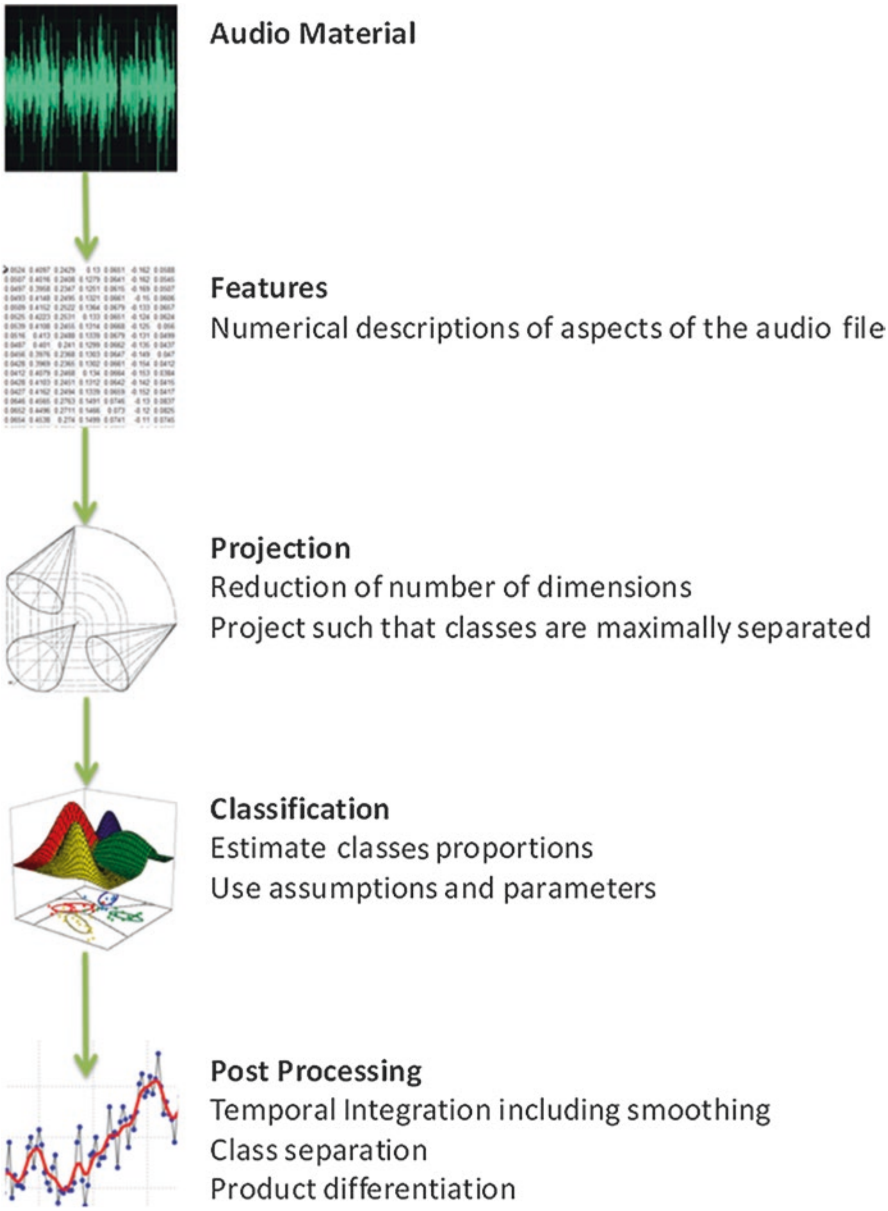

Fig. 4.11 Schematic diagram of how an automatic scene classification algorithm is developed and implemented

The following acoustic parameters can be used for identifying acoustic environments: the degree of synchrony of temporal onsets across different frequency bands; level differences between different frequency bands; the estimated signal-to-noise ratio in different frequency bands; the spectral shape (e.g., the relative energy at high and low frequencies); and the pattern and spectrum of any amplitude modulation. These features are extracted in short time segments (usually corresponding to the frame rate of signal processing in a block-based processing scheme) and evaluated over an observation window that is typically quite long (several seconds), as the objective is to identify changes in the listening environment that typically happen on a long timescale. The information from the environment classification can be used to switch complete listening programs or to adjust the parameters of specific algorithms or combinations of algorithms. Typically, the classification systems are designed to be rather robust and slow to avoid "nervous" (rapid) back and forth switching between different parameter settings.

Some environmental sounds, such as kitchen sounds, shavers, and vacuum cleaners, can last a long time and have acoustic characteristics similar to those of one of the other possible classified environments, such as music. This potentially could lead to inappropriate settings of algorithms. Therefore, a crucial requirement in designing and implementing a robust automatic environment classifier is to train the classifier with a large set of data using real-life scene recordings. The training set should encompass a broad range of types of daily life sounds, including speech samples, music samples, and environmental sounds, together with many different samples of each sound type.

There are some problems and limitations of automatic environment classification systems. One is that the hearing aid user may have different requirements in similar acoustic environments. For example, the requirements of the user are very different if they are sitting in a street café and talking to a friend as opposed to walking along the same street but not having a conversation. In the first case, a directional microphone would be the best choice, whereas in the second case, an omnidirectional microphone would be preferable. To resolve such problems, hearing aids should always offer controls allowing the user to overrule the automatic control system and select his or her preferred listening program.

4.8 Concluding Remarks

Signal processing in hearing aids has come a long way. Over the 20+ years that full digital signal processing has been used in hearing instruments, performance has significantly improved. Hearing aids of today have become intelligent systems that offer processing strategies that are tailored to the individual patient and to specific environments. For ease of operation in everyday life, scene classification algorithms can be used to automatically switch and fade between different processing strategies. Innovation and further developments in technology will be driven not just by the need to improve the intelligibility of speech in background sounds but also by the need to decrease listening effort and to provide good and natural quality for a variety of environmental sounds, including music. Improvements will come from improved signal processing, improved acoustical and mechanical design, including microphone arrays, and more precise fitting of the device to suit individual needs. Challenges and prospects for hearing aid technology are discussed further in Chap. 11 by Popelka and Moore.

Conflict of interest Stefan Launer is an employee of the hearing health care manufacturer Sonova.

Justin A. Zakis declares that he has no conflict of interest.

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Chapter 5
Wireless Technologies and Hearing Aid Connectivity

Jill Mecklenburger and Torben Groth

Abstract A main benefit to end users of wireless features in hearing aids and accessories is the improvement in signal-to-noise ratios when listening to the television and other audio sources in the home as well as when listening to movies, concerts, and lectures in public. Other benefits of wireless hearing aid systems include improved telephone communication and easier control of hearing aid functions using remote controls. These benefits can potentially increase the use of hearing aids among users and in turn contribute to higher customer satisfaction. This chapter examines the history and evolution of wireless and Bluetooth® technology in hearing aids. Signal carriers such as electromagnetic field, infrared, and radio are explained as well as the transmission protocols utilized to deliver the wireless signal. Challenges of wireless transmission in environments with both smaller spaces and larger public spaces are addressed. Power source applications and wireless safety are considered. The most recent advances in wireless interfacing, as well as future directions for wireless hearing aid technology, are discussed.

Keywords Battery • Bluetooth • FM • Frequency modulation • Infrared • Near-field magnetic induction • Radio frequency • Remote microphone • Streamer • Telecoil • Wireless • Zinc-air

5.1 Introduction

Wireless connectivity in hearing aids has been one of the most important advances in the evolution of hearing aid technology. By enabling hearing aids to wirelessly connect directly to sound sources from external devices such as landline telephones, mobile telephones, televisions, stereos, car audio systems, computers and tablets,

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wireless transmission improves the signal-to-noise ratio (SNR) and in turn significantly improves sound quality and the ability to understand speech for the end user. The improvement of SNR is also achieved with wireless transmission to hearing aids from the microphones of public address systems in theaters, concert halls, lecture halls, and places of worship as well as in airports, train stations, and bus stations.

This chapter outlines how wireless transmission to a hearing aid occurs. Variations include transmission from external audio sources at home or at work, hearing aid-to-hearing aid transmission, and transmission in public spaces. Important patient interface factors are discussed, including battery consumption issues and solutions to these challenges.

5.2 Wireless Principles

All wireless transmission systems take the form of a signal carrier and transmission protocol. A signal carrier is the physical medium carrying information from point A to point B. The signal carriers described here are electromagnetic, infrared, and radio frequency (RF). A transmission protocol is a set of rules describing and governing how the physical carrier is used to transmit the signal. In its simplest form, a transmission protocol might define use of the amplitude of frequency modulation of the RF carrier; a more complex transmission protocol would be the seven-layer Bluetooth® protocol describing how digital signal transmission takes place. Each of the following sections describes significant attributes of the carriers and their associated protocols.

5.2.1 Electromagnetic Transmission

The telecoil is a hearing aid internal component consisting of a small magnetic metal core cylinder wrapped in copper wire (Fig. 5.1) that picks up an electromagnetic signal from an external source. Originally intended to pick up stray electromagnetic signals radiating from the loudspeaker of a telephone, this system is probably the earliest version of wireless transmission to hearing aids, dating back to the 1930s. The system evolved over time by intentionally connecting sound sources to an induction loop that creates an electromagnetic field. This can be in the form of a loop of wire around the perimeter of a room for public use or around the neck of hearing aid users for personal use. The orientation of the telecoil cylinder within the hearing aid is important to ensure optimal connectivity as it receives the signal. The ideal telecoil orientation in a hearing aid is dependent upon its relationship to the electromagnetic field of the desired signal. For a looped room, generally a loop of wire around the perimeter of the room in the horizontal plane, the telecoil should be oriented in a vertical position when the hearing aid is worn on the ear (Fig. 5.2). The telecoil produces a voltage when an alternating electromagnetic field flows through it. The telecoil voltage is delivered to the hearing aid amplifier via a separate route

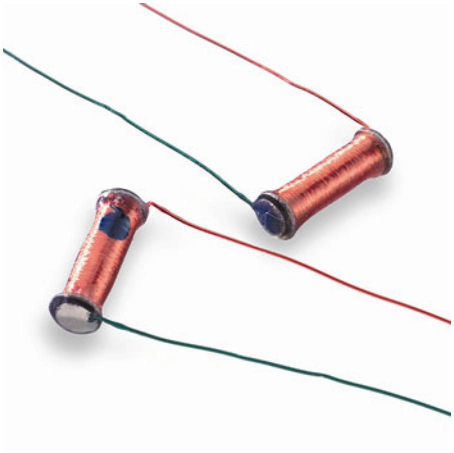

Fig. 5.1 Telecoils (Courtesy of Knowles Electronics)

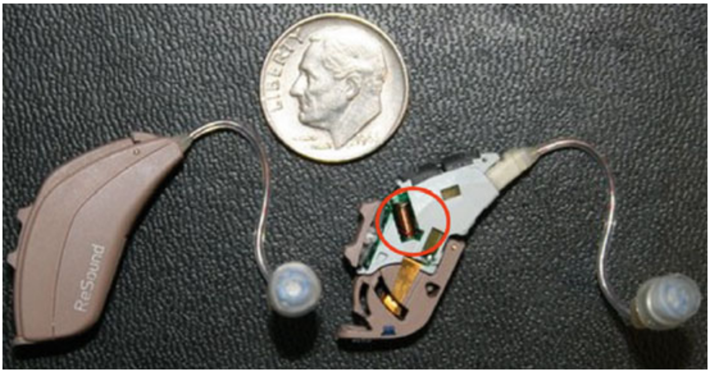

Fig. 5.2 Telecoil orientation within a behind-the-ear hearing aid (Courtesy of GN ReSound)

from that of the hearing aid microphone. The telecoil system is based on direct conversion. As a result, the level of the analog signal picked up by the telecoil is directly proportional to the electromagnetic field intensity and no elaborate decoding is required.

As originally intended, a telecoil can still be used to receive a direct signal from a telephone when the telephone produces a strong magnetic signal and the telephone handset is held in close proximity to the hearing aid. Landline telephones, as well as older mobile telephones, produce the strongest signals. For optimal electromagnetic pickup in this situation, the telephone should be positioned perpendicular to the telecoil. When a home or public venue is equipped with an induction loop, a hearing aid that has a telecoil can receive signals routed through the loop from a television or other audio source in the home or from the audio source in a public space such as a cinema or theater. In lieu of a hardwire loop setup in a home, a

Fig. 5.3 Personal neckloop worn by an individual, including the plug for an audio source (Courtesy of Williams Sound)

Fig. 5.4 International symbol indicating that a location is looped for telecoil reception (www.hearingloop.org)

hearing aid wearer may also wear a loop system around his or her neck that can be plugged into the desired audio source (Fig. 5.3). Public venues equipped with loop systems often display the international symbol for a telecoil loop (Fig. 5.4).

The primary advantages of utilizing a telecoil are the improvement in SNR, the reduced effect of room reverberation, and the general simplicity of the system. Disadvantages include (1) the requirement for installation of the inductive loop; (2) the need for a telecoil within the hearing aid, which increases the size of the hearing aid; (3) the electromagnetic signal from some telephones may be insufficient for reception by the telecoil; (4) the received signal strength may be insufficient if the distance between the loop and the telecoil is too large or if their relative orientations are inappropriate; (5) nonconfidential transmission; and (6) susceptibility to magnetic noise sources.

Although telecoils have been available for decades, they became more popular in the 1960s and 1970s and were commonly used in hearing aids. More recently, telecoils have experienced a comeback as a result of a growing movement in the United States for looping of public venues. In Europe, telecoil use is high within the homes of hearing aid users, many of whom have room loop systems or personal neck loops.

5.2.2 Near-Field Magnetic Induction

Near-field magnetic induction (NFMI) can be thought of as a "personal loop system" that most often incorporates digital technology. It is a magnetic induction system with a short-range wireless system that communicates by coupling a tight, low-power, nonpropagating magnetic field between devices. In contrast to the classic telecoil/loop system, the NFMI system often uses a signal modulation scheme and works more like an RF system, thereby overcoming some of the basic telecoil/loop system shortcomings while also improving the power efficiency. This concept calls for a transmitter coil in one device to modulate a magnetic field that is picked up by a receiver coil in another device and then demodulated. NFMI systems typically operate in the 10–14 MHz frequency band. The modulation scheme is most often digital, offering advantages such as reduced susceptibility to external noise sources and the possibility of increasing security with signal encryption.

NFMI systems are designed to contain the transmission energy within the localized magnetic field, which is shaped almost like a magnetic bubble. The magnetic field energy does not radiate significantly into free space. This type of transmission is referred to as "near field." The power density of near-field transmission rolls off at a rate proportional to the inverse of the range to the sixth power ($1/\text{range}^6$) or −60 dB per decade. This limits the useful range of the NFMI system to 3–5 ft (1–1.5 m).

An important and necessary part of any wireless transmission system is the antenna. For NFMI systems, the antenna consists of a "coil" for transmitting and receiving. This often takes the shape of a neck loop when working as the external part of a hearing aid system, while the antenna in the hearing aid looks more like a conventional telecoil. Transmission between bilateral hearing aids using the internal antenna also has been implemented in commercial products.

The creation of the "magnetic bubble" in combination with the use of a digital transmission protocol makes the NFMI system fairly resistant to interference and disturbances. Another advantage of NFMI is that it can be built to consume relatively little power, which is important when used in hearing aids.

5.2.3 Infrared Transmission

Another way of transmitting wirelessly has been through the use of infrared technology. Although implemented successfully with wireless headphone systems, infrared transmission technology has had very limited success in connection with hearing aids . An infrared wave behaves very much in the same way as visible light: It travels in straight lines and is easily blocked. Although it bounces off flat, light-colored surfaces, the system easily loses connection. Its susceptibility to “noise sources” such as sunlight has further contributed to its limited use in hearing aids.

Wireless headset systems operating with infrared transmission are commonly used in public venues such as theaters and cinemas, as well as on a smaller scale for home use. The advantages of an infrared system include privacy, as there is no spillover of information outside the room of transmission, and the availability of inexpensive systems for home use. A personal infrared listening system for television use can be purchased for less than US$150. Although it is an advantage for some consumers that an infrared system operates independently of a hearing aid, this can be a disadvantage for those wearing hearing aids. Depending on the type of hearing aid, some users may need to remove their hearing aids to wear an infrared receiver. Examples of home/personal infrared systems are shown in Fig. 5.5.

5.2.4 RF Transmission

RF transmission systems provide a more convenient way of getting a signal to a hearing aid user than other systems. This can be attributed to the long-range properties of RF as well as the absence of a need for installations such as loop systems. When using RF transmission, the audio signal is used to modulate the RF carrier according to one of a number of principles. These modulation principles can be analog or digital. For hearing aids , analog frequency modulation (FM) and digital frequency-hopping spread-spectrum (FHSS) are mostly used.

Fig. 5.5 Examples of infrared systems (Sennheiser, *left* and TV Listener, *right*) (Courtesy of Sennheiser)

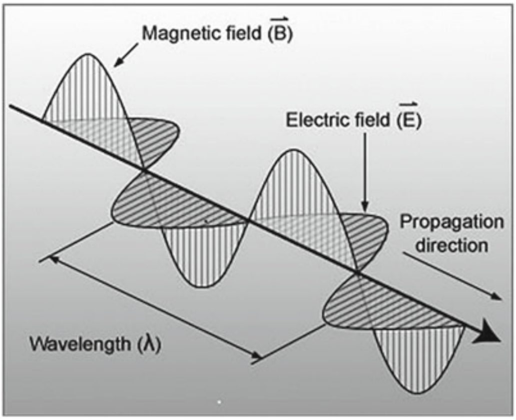

Fig. 5.6 Representation of an electromagnetic wave illustrating the relationship between the accompanying electrical field and magnetic field

The wireless RF system uses an antenna to generate a propagating electromagnetic wave. An electromagnetic wave is made up of an electrical field and a magnetic field (Fig. 5.6). In these types of systems, all of the transmission energy is designed to radiate into free space. This form of transmission is referred to as "far field."

According to Maxwell's equation for a radiating wire, the power density of far-field transmissions rolls off at a rate proportional to the inverse of the range to the second power (1/range^2) or −20 dB per decade. This relatively slow attenuation over distance allows communication over a long range. Typical useful ranges for RF systems when used in personal communication devices such as wireless headsets and hearing aids are 23–30 ft (7–10 m). Figure 5.7 illustrates the difference in obtainable useful range between RF and NFMI carrier technology systems.

Like NFMI systems, RF systems require antennas. Because RF is propagating, there is a fixed relationship between the frequency/wavelength of the transmission and the required physical dimensions of the antenna. A lower frequency means that a longer antenna is required and vice versa. Accordingly, the RF systems in hearing aids today operate at high frequencies to allow a small antenna.

RF systems are more susceptible to interference from other RF signals than NMFI systems. To avoid interference, various schemes allow RF systems to coexist with other transmission systems in the same frequency band. FHSS is a commonly applied example of such a technique. This solution is used in personal communication devices as well as in service provider equipment and military applications. The price paid for the good range is higher power consumption than for NFMI, which is unfortunate in a hearing aid. Advanced transmission protocols can, however, bring the power consumption into an acceptable range. The frequencies of 900, 868, and 2.4 GHz are used by hearing aid manufacturers today for RF wireless transmission in connection with hearing aids. These fall within specific frequency bands called the industrial scientific medical (ISM) bands. Originally reserved internationally for the use of RF energy for industrial, scientific, and medical purposes other than communications,

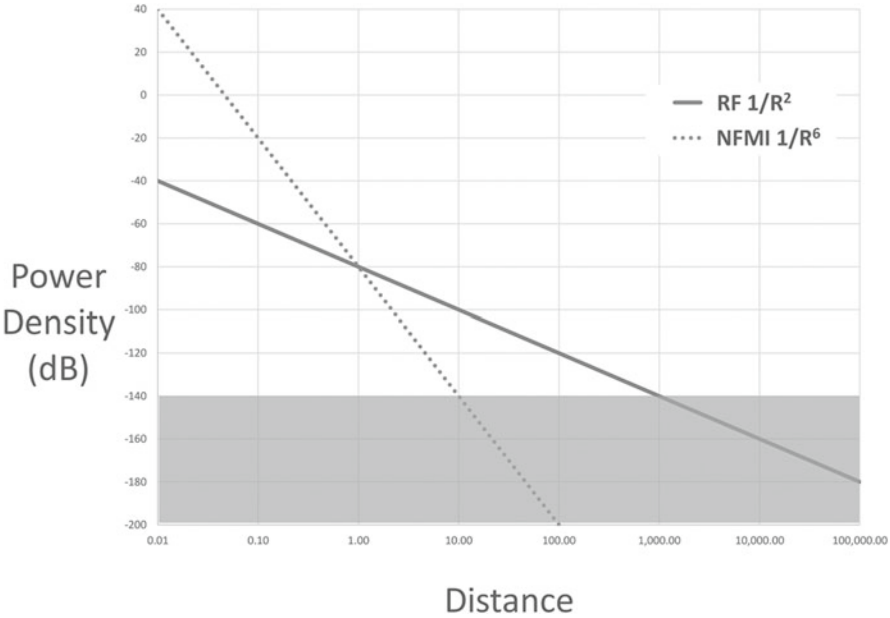

Fig. 5.7 Near-field (NFMI) and far-field (RF) field strengths as a function of distance

the license-free ISM bands can now be used for general public purposes. The 2.4-GHz ISM band has worldwide acceptance. The use of the 900-MHz ISM band is still limited to region 2, which encompasses the Americas, Greenland, and certain eastern Pacific Islands. A good choice for a direct to the hearing aid system is 2.4 GHz because of its good range, excellent data transmission capability, and widespread acceptance. In addition, 2.4 GHz is exclusively used for Bluetooth protocol and company proprietary transmission protocol hearing aid systems.

5.3 Wireless Transmission Systems in Hearing Aids

5.3.1 FM Systems

The legacy industry standard for radio transmission to hearing aids is the FM system. FM systems have excellent sound quality and good range, and therefore FM is used mainly in the school environment as a component of auditory training equipment. Platz (2004) found improvements in SNR of 20 dB or more from an FM system. Other FM system advantages include (1) a long transmission distance, ranging from 50 ft indoors to 100 ft outdoors and (2) the ability to operate on different channels, allowing multiple systems to operate simultaneously in close proximity to one another. Just like FM broadcast radio, these systems require a frequency

Fig. 5.8 FM transmitter (Courtesy of Phonak)

Fig. 5.9 Universal FM receiver with audio shoe (*left*) and integrated FM receiver without audio shoe (*right*) (Courtesy of Phonak)

all to themselves to avoid interference, but a noisy thermostat or electric drill or an FM transmitter on a nearby frequency can all cause electrical interference and/or distortion.

The FM receiver in a hearing aid system can either be a universal component capable of attachment to many behind-the-ear (BTE) instruments or be a proprietary integrated component compatible only with a particular model of BTE. FM systems are relatively easy to install. The disadvantages include high cost, susceptibility to electrical interference, and the requirement to connect additional hardware to the hearing aid, making it larger. Examples of FM transmitters and receivers are shown in Figs. 5.8 and 5.9, respectively.

5.3.2 NFMI Systems

Wireless radios within hearing aids have become the latest method for receiving direct signals from audio sources. This method has evolved from more recent improvements in the miniaturization of RF antennas.

The first generation of this type of wireless hearing aid was based on NFMI technology. In some NFMI systems, the desired audio source—for example, the signal from a television—is connected to a device that digitizes the signal and codes it into the Bluetooth protocol. The device then transmits the signal via 2.4-GHz Bluetooth to a gateway device, called a "streamer," that is typically worn like a lanyard around the neck of the hearing aid user. An example of a gateway device is shown in Fig. 5.10. The signal is then decoded and sent from a radio transmitter coil in the gateway device via a localized magnetic field. Finally, the signal is picked up by a radio receiver coil in a hearing aid and presented as an audio signal to the hearing aid user. The main advantages of NFMI are twofold: ease of implementation because of existing telecoil technology and a low current drain because of the lower carrier frequency of this type of system. Drawbacks include (1) a short transmission distance of approximately 1 m; (2) a requirement that the gateway device be in close proximity to the hearing aids, which is why it is typically worn around the neck; (3) the possibility of compromised sound quality related to the orientation of the gateway device and hearing aid receiver coil; (4) audio delays introduced by the relay time between

Fig. 5.10 Gateway "streaming" device (Courtesy of Widex)

components, causing echo effects and problems with lip synchronization when watching and listening to television; and (5) no possibility of multiple users accessing the same streamer because of the one-to-one relationship between the transmitting device and the streamer.

5.3.3 Bluetooth® Systems

Bluetooth is a wireless generic standardized protocol that allows for transmission of data between fixed and mobile devices. Operating on 79 1-MHz-wide channels in the ISM band (2.4–2.485 GHz), Bluetooth uses FHSS that allows multiple products from the more than 12,000 Bluetooth devices available to work without interfering with each other (Bray and Sturman 2001). The protocols must be flexible enough to accommodate various uses, including wireless networks, mobile headsets, game controllers, and medical equipment. Until very recently, Bluetooth was not a realistic option for direct transmission from an audio source to a hearing aid due to limitations in chip size, Bluetooth audio delay, and high power consumption. Typical audio delays for systems based on the Bluetooth transmission protocol are about 65 ms and in a few instances above 100 ms. Delays of this magnitude are responsible for the echo effects and issues with lip synchronization when, for example, the streamed signal is competing with a direct acoustic signal coming from a TV. Streaming audio to a hearing aid via classic Bluetooth while using a size 10A hearing aid battery would require the battery to be replaced every 2 hours! This example makes it clear that classic Bluetooth is a poor choice for streaming audio to hearing aids.

Bluetooth 4.0 was introduced as part of the main Bluetooth standard in 2010, incorporating Bluetooth Low Energy, which was later termed Bluetooth Smart by the Bluetooth Special Interest Group. Devices operating with this version consume a small fraction of the power of classic Bluetooth and they quickly became available in products within the healthcare, fitness, and home entertainment industries. Bluetooth Smart was aimed at new low-power and low-latency applications for wireless devices within a range up to 50 m. It is because of this Bluetooth version that Made for iPhone® hearing aids became available. This type of direct communication between an iPhone or iPad and a hearing aid is discussed in Sect. 5.6.2.

5.3.4 Company Proprietary RF Systems

Wireless radios in hearing aids utilize transmission in the same ISM bands as the Bluetooth system. The ISM bandwidth is defined and regulated by the International Telecommunications Union, which dictates strict power limitations and requirements for staying in-band. Hearing aids using such systems operate with a proprietary transmission protocol.

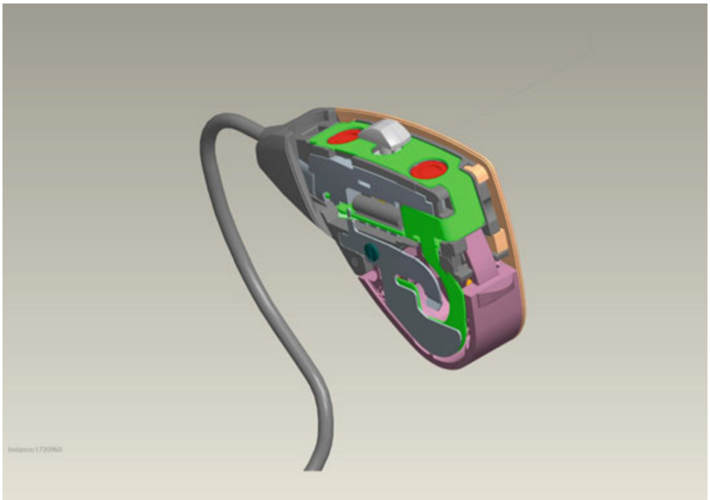

Fig. 5.11 Specially designed RF antenna (*green*) wrapped around hearing aid internal components (Courtesy of GN ReSound)

With proprietary RF technology, the accessories or devices that send audio information to the hearing aid can be further away from the hearing aids than with an NFMI system. Typical distances for an RF accessory or transmitting device can range from 3 m for a two-way communication device such as a remote control to 7 m for a one-way device such as a television streaming device. The distance is determined by transmission power and antenna efficiency. Therefore, this technology does not have an inherent range limitation. The information is sent directly to the hearing aids, without the use of an intermediary or gateway device. Using the same example as described in Sect. 5.3.2 for NFMI transmission, the television signal is digitized and coded by the TV streamer accessory. It is then transmitted via RF to an antenna in the hearing aids where the signal is decoded back into a digitized audio signal (Fig. 5.11). An important concept is that this transmission takes place without the use of a gateway device, thereby simplifying the system.

Current proprietary RF systems use 886- or 900-MHz, or 2.4-GHz ISM bands. Advantages of proprietary RF systems include (1) no intermediary device required for connectivity; (2) low-latency processing from the audio source to the listener, which helps to reduce echo and lip-synchronization problems when watching and listening to television; (3) a signal transmission distance of approximately 7 m; and (4) the ability to stream to multiple instruments from the same transmitter, depending on the manufacturer. The disadvantages of proprietary RF systems are (1) the requirement for a specially designed antenna; (2) inability for some frequencies, such as 886 and 900 MHz, to be used worldwide; and (3) higher battery consumption in the hearing aid than with NFMI systems.

Significant differences exist among the main proprietary RF wireless hearing aid systems that are currently available. While 900- and 886-MHz systems have lower battery consumption than 2.4-GHz systems, an important advantage of a system operating on 2.4 GHz is its ability to be recoded for Bluetooth protocols, given that Bluetooth operates on the same 2.4-GHz ISM band.

5.4 Wireless Connectivity Between Hearing Aids

5.4.1 Hearing Aid-to-Hearing Aid Transmission

Hearing aid-to-hearing aid transmission is also known as ear-to-ear (E2E) transmission or device-to-device (D2D) transmission. By enabling bilaterally fitted hearing aids to communicate with each other wirelessly, several advantages can be obtained, from synchronization of program selection and volume control adjustments to processing of the audio signal.

5.4.1.1 Control Data

Wireless transmission between hearing aids was introduced by Siemens in 2004 for the purpose of ease of use of hearing aid controls. Linking a binaural set of hearing aids wirelessly relieves the end user of the need to reach up to both hearing aids to change volume or listening program if binaural adjustments are desired without the use of a remote control. With wireless control data transmission, volume control or program pushbutton adjustments on either the right or left hearing aid are wirelessly transmitted so that the same adjustment is made simultaneously to the hearing aid on the opposite ear. Activation of external wireless accessories such as streaming from a television or laptop computer can also be linked between the two hearing aids. In an extension of this approach, the control signals of the automatic gain control (AGC) system can be linked across bilaterally fitted aids. This requires a higher data transmission rate, but the rate for control signals is still much lower than for transmission of the audio signal itself. The linking of the two AGC systems ensures that automatic gain changes are synchronized across ears, which may be beneficial in preserving interaural level cues for sound localization (Korhonen et al. 2015). Decision-making algorithms, such as microphone mode switching, can also use information from both hearing aids because of the data exchange possibility.

5.4.1.2 Audio Data

Transmission of the audio signal between two hearing aids allows audio collected from both sides of a wearer's head to be utilized in sound-processing algorithms. For example, a highly directional characteristic can be created, which can be "steered" (either automatically or via a remote control) to focus on sounds to the front, the left, or the right (see Launer, Zakis, and Moore, Chap. 4). Audio data transmission between hearing aids was initially implemented with NFMI. This type of transmission is more challenging to accomplish with a 2.4-GHz system; however, it can be achieved with a special antenna design that optimizes diffraction.

5.4.2 External Device to Hearing Aid Transmission

5.4.2.1 Challenging Environments

Using hearing aids in public venues can be exceptionally challenging, particularly in venues with high levels of background noise. Difficulty understanding speech through hearing aids in noisy restaurants or other social situations has been reported as the most common complaint of hearing aid users as well as the top reason for nonadoption of hearing aids (Kochkin 2007). Wireless connectivity can improve speech comprehension in noisy situations by significantly improving the SNR. The signal from the desired sound source is transmitted directly to the hearing aid, bypassing the hearing aid microphones and allowing for clear reception of the desired sound source.

Listening in Smaller Spaces (Restaurants, Classrooms, Homes, Workplaces)

The FM systems discussed in Sect. 5.3.1 are the standard wireless setups used in most classroom situations. Companion microphone devices utilizing either Bluetooth+NFMI or company proprietary RF for transmission are being increasingly used as a less expensive alternative to the traditional costly FM systems (Christensen 2013). The ease of pairing to hearing aids and the absence of a requirement for extra equipment attached to the hearing aid such as an audio shoe make companion microphones more convenient to use than an FM system. In a noisy restaurant environment these microphones can provide up to 18–20 dB improvement in SNR (Jespersen 2012; Keith and Purdy 2014). Examples of companion microphones are shown in Fig. 5.12.

Listening in Larger Spaces (Auditoriums, Theaters, Places of Worship)

In larger spaces, such as auditoriums and concert halls, the loop systems discussed in Sect. 5.2.1 are the standard systems for improving the SNR (and for reducing the effects of reverberation). Loop systems provide clear sound quality and are a relatively low-cost solution for large-scale wireless transmission simultaneously to all telecoil-equipped hearing aids located within the loop. Specifically, the voice of an actor on a stage can be heard in the back of a very large theater with the same clarity as when standing close to the actor.

Fig. 5.12 Phonak Roger Pen (*top*) and ReSound Multi Mic (*bottom*) (Courtesy of Phonak and GN ReSound)

5.4.2.2 Wireless Accessories for Hearing aids

Telephone

Mobile telephone usage has historically been one of the more demanding challenges for hearing aid users because of problems with limited output level of the stray electromagnetic signals, compromised sound quality, and acoustic feedback. Until recently, wireless transmission from mobile telephones to hearing aids required the use of an intermediary gateway device, both for NFMI systems and for RF systems. For the RF systems, this was solely due to the high power consumption of Bluetooth technology. With the advent of Bluetooth Smart, two manufacturers introduced direct wireless transmission from Apple mobile devices to hearing aids. With direct streaming, any audio source from the Bluetooth Smart iOS device can be transmitted wirelessly in stereo directly to the hearing aids. This was a landmark development in wireless transmission, as it significantly improved mobile telephone compatibility with hearing aids. At the time of writing, it is not possible to stream audio directly (without the use of an intermediary device) from mobile operating systems other than iOS; however, this is an expected development for the future. It is currently possible to transmit control data to and from the hearing aids to both iOS and Android mobile operating systems for remote control functionality and status information.

Although unilateral routing of a telephone signal to a hearing aid (such as with a telecoil) provides benefit, speech recognition and subjective ratings of comfort and ease of use are better with bilateral wireless routing of the telephone signal (Picou and Ricketts 2013).

Personal Music Players

In addition to streaming music directly from Bluetooth Smart iOS devices to hearing aids, there are other methods for wireless coupling of personal music players to hearing aids. Some companion microphone accessories have a line-in feature in which a personal music player (iPod, mp3 player) can be directly connected to the companion microphone with a 3.5-mm connector and cable, enabling direct streaming to the hearing aids.

Television

Television streaming to wireless hearing aids has been shown to significantly improve understanding of the dialogue, particularly in situations with background noise (Sjolander et al. 2009). For the hearing aid user, this alleviates the problem of needing to have the television at a volume that is uncomfortably loud for others who do not have hearing loss. Direct streaming of television audio to a hearing aid also includes the advantage of improving the quality of the signal over amplification solely from the hearing aid. There is evidence that increasing the number of environments in which hearing aid users can better utilize their hearing aids is associated with higher user satisfaction ratings (Kochkin 2005).

5.5 Power Source Considerations

5.5.1 Hearing Aid Battery Requirements

The introduction of wireless functionality to hearing aids has raised the demands on the batteries powering the aids. This is particularly the case when the wireless technology is of the RF type. In addition to powering the audio signal processing and the amplification circuitry, the battery also has to power the wireless RF transmitter/receiver and the processor executing the transmission protocol. Because almost all 2.4-GHz wireless systems are digital and transmit their data in packets, there is an increased demand on the battery to be able to deliver fairly high current "spikes" when transmitting these data packets. For this type of application, batteries with low internal resistance are required when wireless streaming is activated.

With the almost exclusive use of zinc–air batteries, as described in Chap. 1 by Moore and Popelka, it is important to (1) always use fresh batteries, preferably with a shelf life greater than 2 years; (2) use batteries intended for the purpose—for example, batteries marked with "Wireless" might have lower internal resistance, which allows higher peak current capability; and (3) prepare the battery for use by waiting 2 min after the battery seal has been removed to ensure that the chemical process generating the power is fully operational.

5.6 Wireless Interfacing (User Interaction)

5.6.1 Controls on Devices

There are multiple options for activation of wireless technology within a hearing aid. The most basic method is through the use of a pushbutton on the hearing aid. Volume control buttons or toggles can also be utilized for ear-to-ear streaming of control data as described in Sect. 5.4.1.1.

5.6.2 Controls from External Devices (iPhones, iPads, Apps for iPhone and Android)

Hearing aid remote controls provide many benefits to the user, including availability of larger buttons than those on the hearing aid for function operations and discreetness of adjustments. Early forms of remote controls used various methods for transmission, including infrared, ultrasonic, RF, and magnetic induction (Dillon 2001). Current wireless technology allows manufacturers to offer apps compatible with iOS and Android operating systems for control of hearing aid function from the wearer's mobile telephone or tablet. In addition to basic hearing aid controls of program selection and volume and sound quality adjustments, some apps allow for geotagging of preferred settings in specific locations. Geotagging can be beneficial by allowing a user to apply previously selected settings when returning to a frequent location such as a noisy café. This technology continues to evolve with personal electronics that enable users to control their hearing aids from other devices such as the Apple watch.

5.7 Wireless Safety Considerations

5.7.1 Electromagnetic Interference and Electromagnetic Compatibility

Electromagnetic interference (EMI) is an increasing form of environmental pollution. When any wireless technology is applied to a hearing aid, new requirements for allowable environmental interaction—intended as well as unintended—are brought into play (Ott 2009). An example of intended interaction is a wireless local area network (WLAN) transmitting and receiving signals in the same frequency range as a wireless hearing aid as part of its normal operation. Unintended interference could be straying electromagnetic radiation from a microwave oven whose intended use is heating food by vibrating water molecules at their eigenfrequency (2.4 GHz).

Wireless hearing aids are by design able to cope with the aforementioned conditions by following two requirements: (1) the hearing aid must not cause harmful interference and (2) the hearing aid must be able to handle any interference received, including interference that may cause undesired operation.

To ensure that that hearing aids are in compliance with regulations and directives necessary to allow performance under these conditions, they undergo strict testing before they are released for sale. In the United States, for example, hearing aids have to be in compliance with FCC Directive CFR 47 Part 15, subpart C. This directive includes requirements for intended radiators and radiated emission limits for RF devices. Wireless hearing aids must also be in compliance with numerous other regulations that are applicable in other parts of the world.

5.7.2 Wireless User Safety

A few manufacturers have now built 2.4-GHz technology into hearing aids. These devices receive and transmit in the 2.4-GHz band, which is more universally accessible than the FM bands and other common bands used in the past such as 37, 43, 72–76, 173, 183, and 216 MHz, where access depends on country regulations. The power of the radio used is usually far below the levels commonly used in the mobile telephone industry and defined in "EN/ IEC 62311: 2006 assessment of electronic and electrical equipment related to human exposure restrictions for electromagnetic fields (0 Hz–300 GHz)." Wireless hearing aids are tested according to applicable government-mandated testing requirements including those relating to safety and electromagnetic compatibility as well as US FCC requirements for telecommunications. The results of these tests indicate that there is a low likelihood that they will affect other electronic devices and a high likelihood that they will be able to withstand interference from other electronic devices. However, in the spirit of caution, hearing aid manufacturers have issued specific guidelines that are recommended by manufacturers of defibrillators and pacemakers regarding use of mobile devices. For details, hearing aid manufacturers recommend hearing aid users consult the manufacturers of the other electronic devices.

5.8 Wireless Possibilities for the Future

The future of wireless connectivity in hearing aids holds exciting possibilities. Until very recently, hearing aid manufacturers have offered wireless hearing systems on distinctly different wireless platforms, as described earlier. In March 2014, the European Hearing aid Manufacturers Association (EHIMA) announced a new partnership with the Bluetooth Special Interest Group. The main objectives of the partnership include creating a standard for Bluetooth in hearing aids across all manufacturers. When this is accomplished, all devices will operate on similar protocols, enhancing and simplifying the end-user experience. Future directions for connectivity could include direct connections to a desired signal (movie, concert, television) via Bluetooth in both public and private venues. This in turn could increase the utility of hearing aids as well as provide new opportunities for overcoming hearing impairment.

Conflict of interest Jill Mecklenburger declares that she has no conflict of interest.
Torben Groth declares that he has no conflict of interest.

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Chapter 6
Speech Perception and Hearing Aids

Pamela Souza

Abstract Poor speech perception nearly always accompanies sensorineural hearing loss. Although listeners with poorer auditory thresholds experience more difficulty, there is considerable variability in speech perception across individual listeners. Areas of greatest difficulty may include communication in background noise, difficulty understanding talkers with soft voices, hearing speech at a distance, and conversing over the telephone. Some deficits can be easily addressed with hearing aids, while others present challenges. This chapter reviews the effects of hearing loss on speech perception and discusses how hearing aids can compensate for those effects. Topics include patient-specific factors ranging from differences in cochlear damage patterns that affect speech perception to influences of cognitive ability. Environmental factors include the acoustic cues present in the rapidly varying speech signal; the effects of speech spectrum and level, which affect audibility; and effects of background noise and reverberation. The chapter closes with a review of core hearing aid features, focusing on how technology can be used to address issues relevant to speech perception.

Keywords Aging • Audibility • Cochlea • Compression • Digital noise reduction • Directional microphone • Frequency response • Gain • Hearing loss • Listening effort • Reverberation • Spectral resolution • Temporal resolution • Working memory

6.1 Introduction

Hearing loss is a widespread health issue, affecting 10 % of children (Niskar et al. 1998), 20 % of adults, and 50 % of older adults in the United States (Shield 2006; NIDCD 2010). Although the amount of reported difficulty varies, all people with

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hearing loss experience the same problem: poor speech perception. Poor speech perception is the most common reason for people to seek hearing care (Knudsen et al. 2010). Areas of greatest difficulty may include communication in background noise, difficulty understanding talkers with soft voices, hearing speech at a distance, and conversing over the telephone. Except in rare cases, there are no medical or surgical treatments that can improve hearing in cases of sensorineural hearing loss. Consequently, hearing aids or other assistive devices are the most widely distributed treatment to improve speech perception.

The use of hearing aids can decrease hearing handicap by varying amounts depending on the patient and the situation. On average, hearing aid wearers report as much as a 70 % reduction of handicap for speech perception in quiet (compared to a listener with normal hearing in that situation) (Kochkin 2011). However, even in quiet, hearing aids do not eliminate hearing handicap. That limitation contrasts sharply with options for vision rehabilitation, where the most common treatments for vision loss (prescriptive lenses, cataract surgery, and laser vision correction) can nearly eliminate handicap (Kook et al. 2013; Lee 2014). The inability to "correct" hearing loss reflects the complex nature of the compromised auditory system, whereby hearing loss causes auditory deficits beyond simple threshold shifts. Some deficits can be easily addressed with hearing aids, while others present challenges. This chapter reviews the effects of hearing loss on speech perception and discusses how hearing aids can compensate for those effects.

6.2 Patient Factors Influencing Speech Perception

Several investigators have attempted to draw conclusions about cochlear damage patterns from the audiometric configuration. Seminal work by Schukecht (Schuknecht and Gacek 1993; Schuknecht 1994) classified damage patterns in human temporal bones according to the site of lesion, such as hair cell damage, loss of spiral ganglion cells, or damage to stria vascularis. Schuknecht originally proposed that different damage sites would result in different audiometric profiles and potentially in different speech perception abilities. For example, loss of spiral ganglion cells—"neural presbycusis"—was proposed to result in disproportionately poor speech perception. Although one-to-one associations between the cochlear site of lesion and speech perception ability are almost certainly an oversimplification for most human hearing loss, such associations do allow us to consider differences in cochlear damage patterns in relation to differences in speech perception and are likely to be one factor that explains speech perception variability among people with similar audiograms.

Human studies of cochlear damage patterns have been limited by the need to access audiometric data for later-obtained temporal bones. Therefore, most studies in this area have been based on animal models with control over the cause of hearing loss, such as intense noise exposure (Kujawa and Liberman 2006, 2009) or use of ototoxic drugs that modify the biochemical properties of the ear (Schmiedt et al. 2002;

Lang et al. 2003, 2010). Using a novel approach, Dubno and colleagues (2013) surveyed more than 1,700 audiograms and selected the exemplars that fit predefined audiometric ranges derived from animal models of specific damage sites (e.g., metabolic damage linked to stria vascularis vs. sensory damage linked to hair cell survival). Patient history and risk factors were then analyzed for those exemplars. The results were consistent with the ideas put forth by Schuknecht and colleagues. For example, people whose audiograms fit the "sensory" criteria had a significantly higher incidence of noise exposure than those whose audiograms fit the "metabolic" criteria, and the "metabolic" group was significantly older than the "sensory" group.

To illustrate the idea that different underlying damage patterns may lead to different speech perception abilities, Halpin and Rauch (2009) devised a basic illustration of two people with similar pure-tone audiograms but with different underlying damage patterns. In one case, it was assumed that most sensory receptors (inner hair cells and ganglion cells) were present and in the other, that a portion of the basal hair cells were entirely absent. In the first individual, amplification can lead to appropriate frequency-selective information being carried in the auditory nerve and can improve speech perception. In the second individual, who has a "dead region" lacking receptors (Moore et al. 2000), amplification cannot make appropriate frequency-selective information available, and the individual will exhibit a plateau in the performance intensity function. This basic point will come up whenever the relationship between hearing loss, hearing aids, and speech perception is considered: without a means by which all important components of the acoustic signal can be received and transmitted within the auditory system, some degradation of speech perception is inevitable.

To expand this idea in the context of speech perception, consider the schematic representation in Fig. 6.1. The acoustic signal produced by the talker is first subject to the effects of the acoustic environment, including any background noise, reverberation, or a decrease in signal level due to distance between the talker and the listener. Use of a hearing aid or other assistive device further modifies the signal. The resulting signal is received by the listener but must be processed in several stages within the auditory and cognitive systems. At the periphery, the acoustic signal is transformed to a pattern of vibration along the cochlea, which leads to electrochemical processes in the outer and inner hair cells and then to neural encoding via the auditory nerve and its synaptic connections. At the peripheral level, information can be degraded by loss or dysfunction of outer and inner hair cells or by deficits in synaptic transmission. At the neural level, the firing rates of auditory fibers tuned to different frequencies transmit information about a short-term spectrum, changes in spectrum over time, and temporal patterns of amplitude modulation. The detailed timing of nerve spikes (phase locking) may also carry useful information about the temporal fine structure of the sound at each place in the cochlea (Young and Sachs 1979; Moore 2014). Reliance on that transmitted information has downstream effects on speech perception. For example, hearing loss is thought to shift the encoding balance of envelope and temporal fine structure (Kale and Heinz 2010; Scheidt et al. 2010; Swaminathan and Heinz 2011), a change that may have consequences for the ability to perceive speech in modulated backgrounds.

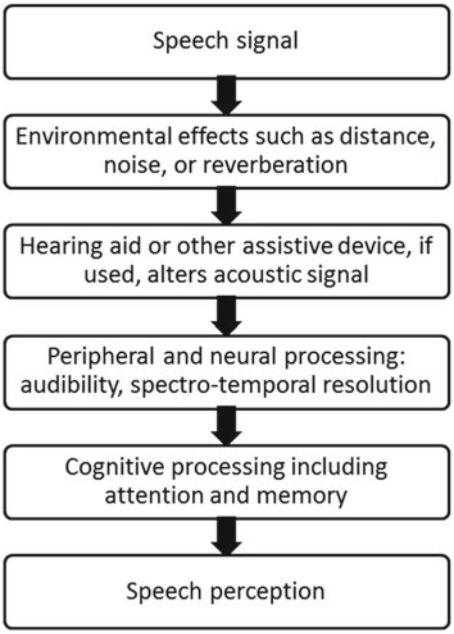

Fig. 6.1 Schematic of stages in the transmission and processing of speech, each of which can affect speech perception

At a later stage, information received via the auditory pathway is subjected to cognitive processes that compare information in working memory with long-term knowledge of phonology, syntax, and semantics to construct the meaning of the signal (Ronnberg et al. 2013). A disruption anywhere in this complex, multilevel process could potentially result in a deficit in speech perception.

To summarize, speech perception is influenced by many factors, including the acoustic environment; any enhancement or distortion of the acoustic information produced by a hearing aid; the processing capabilities of the listener's peripheral and central auditory systems; and the listener's cognitive abilities. Sections 6.3–6.5 consider the contributions to speech perception of each of the last three factors.

Table 6.1 provides a framework for relating possible auditory damage patterns to degree of hearing loss as measured using the audiogram, along with options for treatment with a hearing aid or cochlear implant. With regard to acquired hearing loss via exposure to noise or ototoxic agents, outer hair cells are likely to be the most susceptible, although loss of synapses and auditory neurons may also occur, and inner hair cells may be damaged by impulsive sounds such as gunshots. Although the initial mechanism of age-related hearing loss may be metabolic (specifically, changes to the endocochlear potential) (Schuknecht 1994; Lang et al. 2003, 2010; Saremi and Stenfelt 2013), changes to the endocochlear potential affect both inner and outer hair cells. Therefore, the effect on auditory thresholds is likely to be similar to direct outer hair cell damage from other causes. Some auditory models indicate that complete loss of the cochlear amplifier associated with outer hair cells will result in 50–60 dB of threshold elevation (Ryan and Dallos 1975;

Table 6.1 Expected cochlear damage for different amounts of hearing loss as measured using the audiogram

Degree of loss	Expected cochlear damage pattern	Rehabilitation options
Normal	Intact hair cells; cannot rule out degeneration of synapses and neurons	No hearing aid needed
Mild	Primarily loss of outer hair cells but may also be loss of synapses and neurons	May require hearing aid if objective or perceived communication is affected
Moderate	Loss of outer hair cells; some loss of inner hair cells/synapses/neurons	Partial audibility of conversational speech; hearing aid is recommended
Moderately severe	Loss of outer and inner hair cells and/or synapses/neurons	Poor audibility of conversational speech; hearing aid is recommended
Severe	Substantial inner hair cell loss and probable dead regions	Inaudibility of conversational speech; hearing aid is essential, but benefit may be restricted by poor auditory resolution. A cochlear implant may be considered
Profound	Substantial inner hair cell loss and probable dead regions	

Cheatham and Dallos 2000). In terms of speech perception, one consequence of outer hair cell loss is reduced frequency selectivity (broader tuning), which is discussed in Sect. 6.4.2. This reduces the number of independent channels that can code information about the signal envelope, further impairing speech perception in noise (Swaminathan and Heinz 2011). Speech perception in noise may also be impaired by collateral degeneration of spiral ganglion nerve fibers after the more immediate damage to outer hair cells (Kujawa and Liberman 2006, 2009).

For greater degrees of hearing loss, loss of both outer and inner hair cells is expected (Stebbins et al. 1979; Hamernik et al. 1989; Nelson and Hinojosa 2006). Whereas loss of outer hair cells elevates the tips of neural tuning curves but not their tails, a combined loss of inner and outer hair cells shifts both the tips and tails of tuning curves to higher levels (Liberman and Dodds 1984), significantly affecting the transmission of auditory information. Some information may not be transmitted at all in cases of areas of missing or very sparse inner hair cells, termed "dead regions" (Moore 2004). Dead regions are often associated with severe hearing loss and often lead to perceived distortion of sounds (e.g., Huss and Moore 2005), poor sound quality, and reduced benefit from amplification.

6.3 Audibility

A prerequisite for speech perception is audibility. Speech sounds that fall below the auditory threshold cannot be perceived. For a sentence spoken at a constant vocal level, the level measured in narrow bands (typically, 1/3 octave bands) using

125-ms time windows varies by as much as 50 dB (Dunn and White 1940; Cox et al. 1988), although audibility of the full range may not be necessary for perception (Studebaker and Sherbecoe 2002; Moore et al. 2008). The range of levels is increased further by changes in overall level produced by variations in talker-listener difference and speaking effort. This concept is illustrated in Fig. 6.2, which represents levels in the ear canal for weak, medium, and intense speech presented to a listener with a severe hearing loss while wearing a hearing aid. In each panel, the range of speech levels (enclosed by dashed lines) is plotted relative to the listener's hearing thresholds (filled circles). For the weak (50 dB SPL) input level, only a

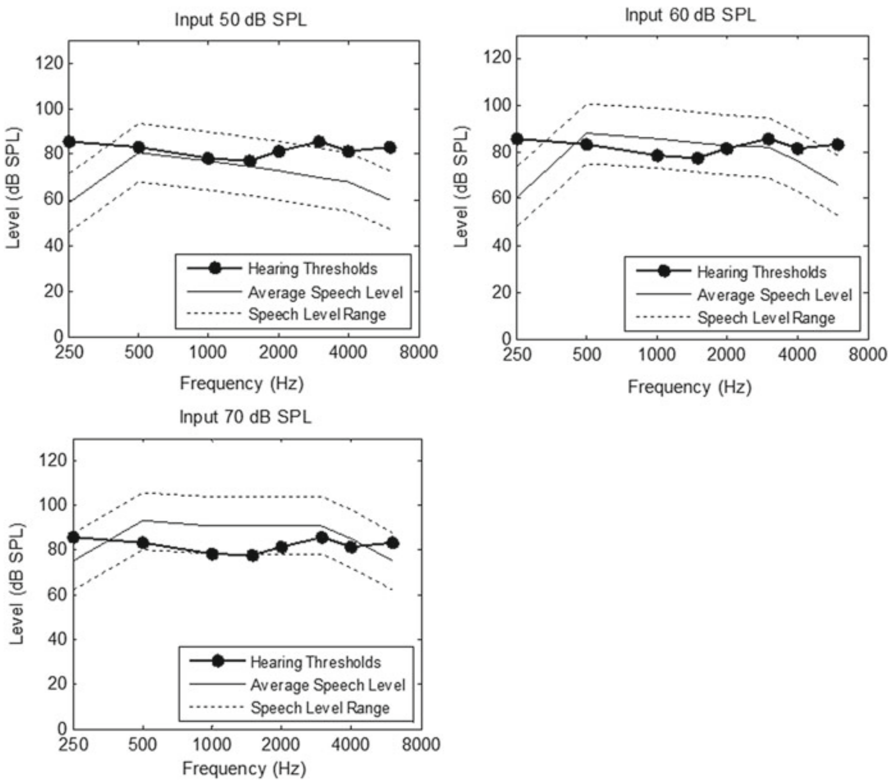

Fig. 6.2 A simple representation of the audibility of amplified speech (1/3 octave bands) for a listener with a severe hearing loss, wearing a hearing aid. The lines without symbols show the short-term range of speech levels (*dashed lines*) about the long-term average level (*solid line*); levels were measured in 125-ms windows. Each panel represents a different speech input level. In each panel, the audible part of the speech range is the area below the top *dashed line* (which represents the most intense speech segments) and above the *thick line* and *filled circles* (which represent the listener's hearing thresholds). For the lowest input level of 50 dB SPL, even with hearing aid amplification, only 23 % of the speech information is audible. For the medium input level of 60 dB SPL, 52 % of the speech information is audible. For the highest speech level of 70 dB SPL, 76 % of the speech information is audible

small portion of the speech information is audible. More generally, audibility is greater in cases of higher speech levels or lower threshold levels and lower in cases of weaker speech levels, higher threshold levels, or the presence of masking noise (not shown in the figure).

Speech perception is determined, in part, by how much of the speech *intensity* range is audible but also by how much of the speech *frequency* range is audible. A classic measure that takes intensity and frequency into account is the Speech Intelligibility Index (SII; ANSI 1997) and its precursor, the Articulation Index (ANSI 1969). The SII is a measure of audibility ranging from 0 (inaudible) to 1 (audible) and is calculated from the proportion of the signal that is audible in each frequency band. The calculation takes into account the importance of each frequency band to speech perception. Audibility depends on the characteristics of the listener (auditory thresholds), the spectrum of the signal, and the spectrum of any background noise. Effects of reverberation are not taken into account. It is important to note that the SII value is not the predicted speech perception score; rather, a transfer function must be used to relate the SII value to intelligibility (Studebaker and Sherbecoe 1991; Souza and Turner 1999; McCreery and Stelmachowicz 2011). For listeners with normal hearing, and presumed good frequency resolution, speech intelligibility is well predicted by audibility (Dubno et al. 1989b). However, for listeners with hearing loss, speech perception is more variable and often falls below that predicted from audibility (Souza et al. 2007). This may be particularly true for listeners with greater amounts of hearing loss, especially listeners with dead regions (Baer et al. 2002; Malicka et al. 2013). The shortfall has been attributed to poor resolution and transmission of acoustic information. Some SII models incorporate a "proficiency factor" to capture these differences (Scollie 2008). Figure 6.3 illustrates this concept using data from a group of 27 listeners, with ages from 70 to 90 years. The figure shows the speech reception threshold (speech-to-noise ratio [SNR], required for 50 % correct) as a function of amount of hearing loss (three-

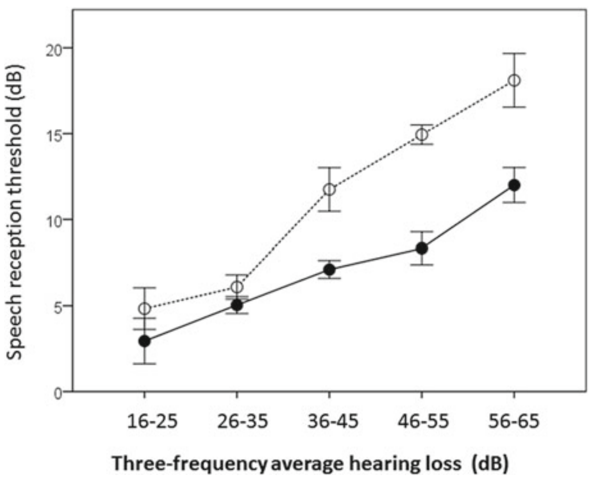

Fig. 6.3 Speech reception threshold (dB SNR at threshold) as a function of three-frequency (0.5, 1, 2 kHz) pure-tone average. A larger *y*-axis value indicates poorer speech reception. *Open circles* show unaided performance and *filled circles* show performance while wearing appropriately fitted hearing aids. In both conditions, greater amounts of hearing loss are associated with poorer speech reception

frequency [0.5, 1, 2 kHz] pure-tone average). A higher speech reception threshold (SRT) indicates poorer speech reception, that is, the listener required a more favorable SNR to understand the speech. Open circles show unaided performance and filled circles show performance while wearing appropriately fitted hearing aids. In both conditions, greater amounts of hearing loss are associated with poorer speech reception. The effect of poor hearing is greater for the unaided condition, where both reduced audibility and poor auditory analysis would be expected to play a role in determining performance. For the aided condition, where audibility is expected to be increased and hence to have less influence, speech reception still worsens with increasing hearing loss, probably because of progressively poorer auditory analysis of audible signals. This is discussed more fully in Sect. 6.4.

6.4 Suprathreshold Resolution

Speech signals vary rapidly in intensity and in spectrum, as illustrated in Fig. 6.4. The top panel shows the waveform and the bottom panel shows a narrowband spectrogram for the sentence "The lazy cow lay in the cool grass," spoken by a female talker and sampled at 22.05 kHz. The figure shows the variation in short-term level over time (along the *x*-axis) and frequency (along the *y*-axis), with higher energy shown as darker shading. To analyze this stream of acoustic information, listeners with normal hearing have the advantage of fine resolution in both the spectral and temporal domains. For example, a listener with normal hearing can detect

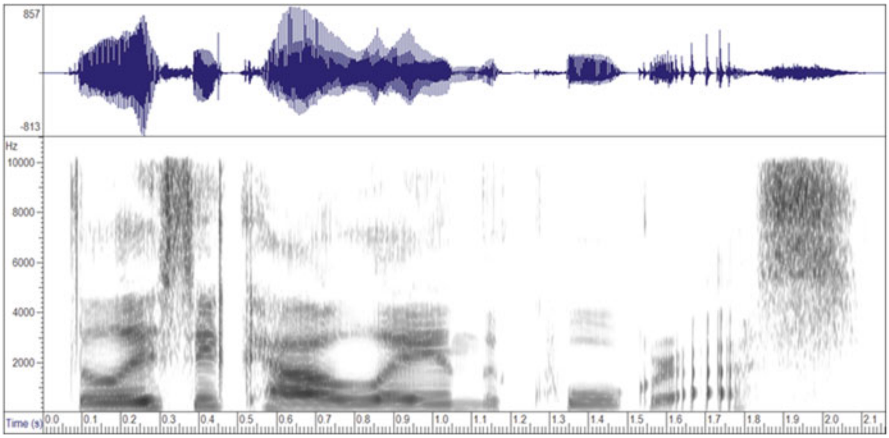

Fig. 6.4 Two representations of the sentence "The lazy cow lay in the cool grass." The *top panel* shows the waveform, that is, the instantaneous amplitude as a function of time. The *lower panel* shows the spectrogram, with frequency on the *y*-axis and higher amplitudes represented by *darker shading*. The figure illustrates the rapidly changing and complex nature of the speech signal. Note, for example, the difference between the dynamic, lower frequency energy of the vowel formant transitions at 0.1–0.3 s (diphthong /aI/ in "lazy") and the static fricative energy at 1.9–2.1 s (/s/ in "grass")

frequency differences as small as a few hertz and detect variations in energy over a few milliseconds (Fitzgibbons and Wightman 1982; Moore 1985). Those abilities allow easy translation of the spectrotemporal variations in the speech signal into meaningful sound.

Once damage occurs to the cochlea or other auditory structures, suprathreshold resolution is often reduced, sometimes in unpredictable ways. Although greater amounts of sensorineural hearing loss are commonly associated with degraded resolution (and therefore poorer speech perception), it is difficult to predict resolution abilities for a particular listener based on their audiogram. Sections 6.4.1 and 6.4.2 review some effects of hearing loss on spectral and temporal resolution and implications for speech perception.

6.4.1 Temporal Resolution

For convenience, speech features can be categorized according to their dominant fluctuation rates. One approach is to consider three rate categorizations: slow (envelope, 2–50 Hz), medium (periodicity, 50–500 Hz), and fast (fine structure, 500–10,000 Hz) (Rosen 1992). Other researchers have proposed that envelope and fine structure should be described in terms of the processing that occurs in the cochlea and that the rapidity of envelope and temporal fine structure fluctuations depends on the characteristic frequency within the cochlea (Moore 2014). Regardless of the nomenclature, we know that different fluctuation rates make different contributions to speech perception. For example, prosodic cues are partly conveyed by slowly varying envelope, whereas segmental cues such as consonant place may be partly conveyed by rapidly varying fine structure. In addition, the relative contribution of each type of information depends on the situation. For example, listeners with normal hearing can perceive nearly 100 % of speech in quiet when that speech is processed to preserve envelope cues but disrupt temporal fine structure cues (Shannon et al. 1995; Friesen et al. 2001; Souza and Rosen 2009). Temporal fine structure is thought to be important for listening in background noise (Moore 2008) as well as for music perception (Heng et al. 2011).

Poor temporal resolution for listeners with hearing loss (compared to listeners with normal hearing) is thought to be related to reduced sensation level and/or narrower stimulus bandwidth (Reed et al. 2009). However, many listeners with hearing loss are older, and age may introduce different problems with temporal processing. Consider how the temporal fine structure of a signal is conveyed through the auditory system. The frequency of a tone is represented, in part, by the time intervals between nerve "spikes." In a normally functioning system, the interspike intervals are close to integer multiples of the period of the tone. With increasing age, the neural firing patterns may become disorganized such that they fail to faithfully represent the signal frequency. Some authors have proposed that this neural disorganization, or "dyssynchrony," will impair the representation of sound at the level of the auditory brainstem (Pichora-Fuller et al. 2007; Anderson et al. 2012; Clinard and Tremblay 2013). Those listeners with poor neural representation also demonstrate poor objective (Anderson et al. 2010, 2011) and subjective (Anderson et al. 2013a) speech perception in noise.

In summary, the ability to resolve some types of temporal information (such as envelope information) may be relatively well preserved in people with sensorineural hearing loss. Other aspects of temporal information (such as temporal fine structure) are likely to be degraded by age and/or hearing loss. However, the extent to which temporal cues are preserved depends on the specific cue under study, the degree of hearing loss, the age of the listener, and perhaps other factors (such as hearing loss etiology) that are not yet well understood.

6.4.2 Spectral Resolution

Excluding conductive pathology, it is expected that most naturally occurring hearing loss involves some loss of outer hair cells. The consequences of outer hair cell loss are reduced audibility (caused by reduced gain of the cochlear amplifier) and reduced frequency selectivity. Listeners with cochlear hearing loss have broader-than-normal auditory filters (Glasberg and Moore 1986). The extent of the degradation roughly follows the degree of loss, so listeners with severe-to-profound sensorineural loss are likely to have very poor frequency selectivity. However, there can be large variability from person to person (Faulkner et al. 1990; Souza et al. 2012b). Degraded frequency selectivity is likely to be one of the major factors affecting speech perception. For speech in quiet, poor frequency resolution impedes accurate representation of spectral shape (Dubno et al. 1989a; Souza et al. 2012b, 2015). For speech in noise, masking effects are increased, causing the noise to obscure spectral features of the target speech (Leek et al. 1987; Leek and Summers 1996). A similar effect can be simulated for listeners with normal hearing by spectral "smearing" (Baer and Moore 1993).

6.5 "Top-Down" (Cognitive) Processing Ability and Listening Effort

Audiological care is usually focused on the capabilities of the peripheral auditory system. For example, clinical evaluations are based on the pure-tone audiogram, which provides information about audibility. Tests for dead regions have been suggested for use in selection of the hearing aid frequency response (Moore and Malicka 2013). The most common clinical speech perception test is monosyllabic word recognition in quiet, although tests of speech perception in noise are beginning to gain traction (Taylor 2003). Specific measures of suprathreshold resolution are infrequently included (Musiek et al. 2005), and it is unclear how those measures should be taken into account when making rehabilitation choices (Sirow and Souza 2013). Although there is considerable interest among clinicians, contributions of the cognitive system to speech perception are not usually assessed or considered as a routine part of audiological care. However, consider the demands of everyday communication: the listener must process a rapidly varying stream of acoustic information; match that acoustic information to stored lexical information to obtain meaning; and retain the information for later access and comparison with new information. It seems reasonable to expect that, in most situations, speech perception will depend on cognitive abilities, including memory and attention, and that those abilities will also affect the ability to understand and remember speech.

Recent work on speech perception and cognitive ability has focused on working memory, which refers to the ability to process and store information while performing a task (Daneman and Carpenter 1980; Baddeley 2000). Ronnberg et al. (2013) postulate that working memory involves deliberate and effortful processing, especially when the auditory representation of the input signal is degraded by noise, by a hearing aid, or by impaired processing in the auditory system. In that view, working memory plays only a minor role in the perception of speech in quiet or when contextual information is available to support a lexical decision (Cox and Xu 2010). Behavioral and physiological data support the idea that adults with poor working memory have poorer speech perception in complex listening environments (Akeroyd 2008; Wong et al. 2009). Such adults also report greater communication difficulty than listeners with similar amounts of hearing loss but better working memory (Zekveld et al. 2013). Because low working memory is associated with poor perception of acoustically degraded signals, it may also affect how an individual responds to signal-processing manipulations in hearing aids (Lunner and Sundewall-Thoren 2007; Arehart et al. 2013a).

Traditional studies of speech perception typically used percent correct or SRTs to compare results across individuals or groups. When there was no difference in score, it was assumed there was no difference in speech perception ability. However, such comparisons do not account for situations where one listener might apply greater conscious or unconscious effort to achieve the same level of speech perception as another listener. As one example, consider a simple intelligibility task (Wong et al. 2009) where older and younger listeners were asked to identify words in multitalker babble at 20 dB SNR (a relatively easy task). Although speech perception scores were similar for the two age groups, functional magnetic resonance imaging (fMRI) results showed reduced activation in the auditory cortex and an increase in working memory and attention-related cortical areas for the older listeners. In other words, equal performance was achieved only by the older listeners expending more cognitive effort to compensate for deficits in auditory and cognitive processing. Effort has also been shown to be correlated with working memory; people with lower working memory expend greater effort (Desjardins and Doherty 2013).

6.6 Language Experience and Effects of Age

A detailed consideration of the effects of age on speech perception is beyond the scope of this chapter. However, speech perception, by its nature, depends on language experience. Experience is one factor that may modify speech perception for younger or older listeners. For children, speech perception skills require time to mature (Hnath-Chisolm et al. 1998; Eisenberg 2007; Werner 2007). The last skills to develop involve speech perception in difficult listening environments, including background noise (Leibold and Buss 2013; Baker et al. 2014). As for adults, listening in these environments may require children with hearing impairment to use cognition to compensate for degraded auditory perception (Osman and Sullivan 2014).

Most hearing loss occurs gradually due to aging, noise exposure, or other late-occurring etiologies. The loss usually occurs in the context of long language experience, and language experience confers some protection against loss of auditory information. For example, older listeners appear to be better able than younger listeners to use context to fill in missing information (Lash et al. 2013). Note, though, that use of contextual information to compensate for degraded auditory input requires deployment of cognitive resources (Aydelott et al. 2011). Accordingly, older listeners' ability to use contextual information may also depend on their cognitive abilities, including working memory (Janse and Jesse 2014).

Overall, there is little doubt that older listeners have more difficulty perceiving speech than younger listeners with similar levels of hearing loss (Gordon-Salant et al. 2010). These deficits are most obvious in complex listening environments (Pichora-Fuller and Souza 2003). Poorer performance in background noise and with rapidly varying signals, such as time-compressed speech (Jenstad and Souza 2007), may be related to degraded neural representations of temporal information (Anderson et al. 2011). Language or listening experience may partially offset those effects (Anderson et al. 2013b) and provide the ability to compensate for peripheral and central deficits.

6.7 Situational Factors Influencing Speech Perception

6.7.1 Background Noise

The most common complaint of people with hearing loss (and sometimes of people with normal hearing!) is difficulty listening in background noise. Most everyday situations involve some level of noise, ranging from favorable SNRs in relatively quiet situations (such as the listener's home or workplace) to negative SNRs in restaurants or public transportation (Olsen 1998). The more spectral, temporal, or spatial "overlap" there is between the talker and background, the more difficult is speech perception. For example, a distant engine is unlikely to interfere with understanding a talker who is situated close to the listener because the engine noise is distinct in frequency spectrum, temporal pattern, and location from the talker's voice. In contrast, attending to a talker in the presence of a second, unwanted talker standing next to the first talker is more challenging. In that case, the target and masking talkers may be producing sound that has similar frequency spectrum, temporal patterns, and location. The listener may need to expend more effort to focus on the target talker. The extent to which a noise "masks" (interferes with) perception of a target depends on a number of acoustic features of the two signals, including similarity of modulation patterns (Stone et al. 2012). Sections 6.7.1.1 and 6.7.1.2 consider the effect of noise on speech perception in two broad categories, energetic/modulation and informational masking.

6.7.1.1 Energetic and Modulation Masking

Energetic masking occurs when the peripheral response to the signal-plus-masker is almost the same as the response to the masker alone (Brungart et al. 2006). Energetic masking is reduced when there is a difference in the peripheral response to the signal-plus-masker and to the masker alone. Such a difference might occur because there is little overlap between the spectra of the target and masker (as for a target talker with a low-frequency voice speaking in the presence of a high-frequency fan), or because of brief reductions in the level of the masker. The noise encountered in everyday listening rarely has a constant level. Moreover, amplitude modulations can occur at different time points in different frequency regions. Listening in spectro-temporal "dips" in the background can decrease energetic masking and improve speech perception. Listeners with normal hearing can take advantage of momentary dips in the background where the SNR is briefly improved to obtain information about the target speech (Festen and Plomp 1990). When the background is speech, the amount of amplitude modulation is considerable when there are only a few talkers but decreases as the number of background talkers increases (Simpson and Cooke 2005; Rosen et al. 2013). Based on the principles of energetic masking, the most effective masker should be a broadband noise with a spectrum shaped to that of the target speech because such a noise does not have pronounced temporal or spectral dips. In practice, this may not be the case, for reasons explained below in the next paragraph.

Stone et al. (2012) have proposed that speech perception is better for speech in modulated noise than for speech in steady noise not because of release from energetic masking but because of release from modulation masking. Modulation masking occurs when amplitude fluctuations in the background make it harder to detect and discriminate amplitude fluctuations in the target signal (Bacon and Grantham 1989; Houtgast 1989). When the background is "steady" noise, random amplitude fluctuations in the noise produce modulation masking of the target speech. When the background sound contains pronounced spectrotemporal dips (over and above those associated with the random inherent fluctuations in the noise), these provide "clean" glimpses of the target speech, free from modulation masking, and this leads to better speech intelligibility. In that view, masker modulation can either increase or decrease speech intelligibility depending on the masker properties. Regardless of the mechanism, there is strong evidence that listeners with hearing loss have impaired glimpsing ability (Takahashi and Bacon 1992; Dubno et al. 2003; Wilson et al. 2010). Possible causes include reduced audibility of the target speech in the masker gaps (Bernstein and Grant 2009) as well as the limitations in auditory analysis described earlier in this section. For example, poor frequency selectivity may limit the ability to glimpse in a narrow spectral dip, and greater susceptibility to forward masking at low sensation levels may limit the ability to glimpse in a brief temporal dip (Festen and Plomp 1990; Gustafsson and Arlinger 1994; Eisenberg et al. 1995).

6.7.1.2 Informational Masking

Informational masking occurs when the listener cannot distinguish the target from the background, even when energetic or modulation masking is not the cause. This happens when the target and masker are confusable and/or similar—as when two people talk at the same time. Informational masking occurs for listeners with normal hearing and with hearing loss (Kidd et al. 2002; Alexander and Lutfi 2004). Because the "noise" in many occupational or social environments includes other talkers, informational masking plays a significant role in everyday listening. Informational masking can also occur when the masker is not speech but is acoustically similar to speech (Souza and Turner 1994; Brungart 2001). For example, informational masking can occur when the masker is a language not understood by the listener (Garcia Lecumberri and Cooke 2006; Van Engen and Bradlow 2007) or when the masker is speech modified to be unintelligible (Freyman et al. 2001; Hoen et al. 2007; Cullington and Zeng 2008). Although some studies have suggested that informational masking may be greater for older listeners or for listeners with hearing loss (Kidd et al. 2002), others have not (e.g., Souza and Turner 1994; Rothpletz et al. 2012).

6.7.2 Reverberation

When listening to speech in a room, part of the speech energy arrives directly at the ears. Other speech energy reaches the ears after reflections from surrounding surfaces, and this energy is delayed relative to the direct signal. The amount of this *reverberation* is often defined by the reverberation time, RT_{60}, which is the time that it takes for the reflections to decay by 60 dB. Reverberation reduces amplitude modulation depth and can affect speech perception in two ways: overlap and self-masking (Nabelek et al. 1989). Overlap masking occurs when reflections from one speech sound overlap in time with a following sound. As a result, whereas noise causes more errors in identification of initial consonants in words, reverberation causes more errors in identification of final consonants (Helfer 1994). Self-masking refers to the distortion of the spectrotemporal information within a single speech sound, such as disruption of formants within a diphthong (Nabelek 1988).

As RT_{60} increases, speech perception worsens (Duquesnoy and Plomp 1980; Shi and Doherty 2008). Listeners with hearing loss may be especially sensitive to distortion from reverberation (Helfer and Huntley 1991; Sato et al. 2007). One source of this problem may be that listeners with hearing loss depend to a greater degree on temporal cues, and these are distorted by reverberation (Nabelek et al. 1989). Unlike listeners with normal hearing, those with hearing loss may be unable to adjust perception to listen effectively in reverberant environments, perhaps because they cannot perceive the acoustic information that is necessary to support that adjustment.

6.8 What to Expect from the Hearing Aid

Hearing loss—and the impaired speech perception that results—has enormous consequences for communication. With very few exceptions, the only treatment available to improve speech perception is amplification, usually via hearing aids, and sometimes via other assistive listening devices. Hearing aids have an onerous task: to improve speech audibility and to preserve essential speech cues while avoiding distortion. Considering the difficulties, hearing aids are effective at improving speech recognition in many situations, particularly in quiet environments. However, they may provide limited benefit in difficult listening environments, including distant talkers where the talker's voice fails to reach the hearing aid microphone at a sufficiently high level; noisy rooms; and highly reverberant situations.

In this section, the focus is how technology might be used to address the issues relevant to speech perception covered in this chapter. Although a detailed review of hearing aid processing is provided elsewhere in this volume (Killion, Van Halteren, Stenfelt, and Warren, Chap. 3; Mecklenburger and Groth, Chap. 5), core hearing aid features are considered in relation to their effect on speech perception. This section also considers the relationship between hearing aid processing and the listener's cognitive ability.

6.8.1 Overcoming Audibility Loss

The basic role of hearing aids is to improve audibility. Listeners with hearing loss whose dynamic range (from threshold of audibility to threshold of discomfort) is less the dynamic range of speech will be at a disadvantage if linear amplification is used, in that either low-intensity sounds will be inaudible or high-intensity sounds will be uncomfortably loud. Focal loss of inner hair cells may also have implications for hearing aid use because it may not be possible to improve reception of signal components falling within the frequency range of the dead region (Hogan and Turner 1998; Vickers et al. 2001; Baer et al. 2002). Fortunately, several types of hearing aid processing can be used to address this issue.

6.8.1.1 Frequency Gain Shaping

Hearing aids are usually fitted in such a way that frequency bands for which hearing threshold is poorer (and audibility is lower) receive greater gain. Over the past 50 years, many schemes have been proposed that prescribe gain at each frequency

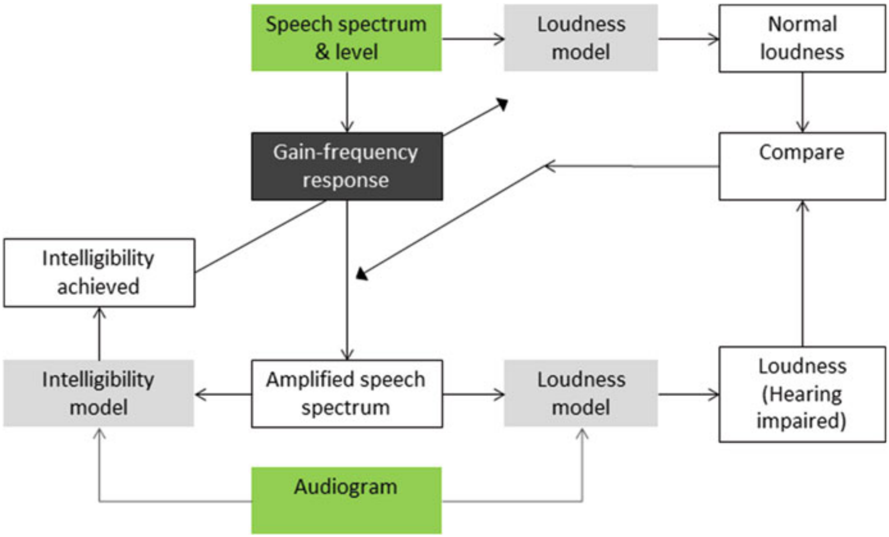

Fig. 6.5 Illustration of the adaptive process used to derive a frequency-gain response that takes into account the audiogram of the listener and the input signal [Modified from Dillon et al. (2011) with permission of the author]

based on the audiogram or sometimes on measures of loudness perception (Byrne and Dillon 1986; Cox 1995; Moore et al. 2010). Although all of the methods were based on sound theoretical principles, only some gained widespread acceptance. Some were abandoned when they lacked updated versions that accommodated new amplification technology; some because they received little validation; and some were inconvenient to implement in clinical practice. Here, three procedures in current use are described as illustrations of the process by which speech perception can be improved via improved audibility.

The first procedure, the National Acoustic Laboratories nonlinear procedure (NAL-NL2; Dillon et al. 2011), aims to maximize speech intelligibility while keeping the overall loudness of the signal at or below that for a normal-hearing listener presented with unamplified speech. The target frequency- and level-dependent gains are derived from a modified version of the SII and a model of loudness perception (Moore and Glasberg 2004) (Fig. 6.5). Frequencies that do not contribute to higher SII values receive little or no gain. The prescription includes a modification for tonal languages, which are likely to differ in frequency content (and therefore require different audibility criteria) compared to nontonal languages. Currently, this is the most common procedure used to fit hearing aids for adults in the United States and in Australia.

The underlying tenet of the second procedure, the Desired Sensation Level procedure and its latest implementation, DSL v5 (Scollie et al. 2005; Moodie et al. 2007), is that audibility will be beneficial. To that end, DSL prescriptions often result in a wider audible bandwidth, greater gain, and sometimes higher compression ratios (discussed in Sect. 6.8.1.3) than NAL-NL2. In the United States and Canada, DSL is a popular choice for pediatric hearing aid fitting, due to its attention to child-specific speech spectra, such as the differences between a listener who is facing the talker and a small child or infant being held by a talker (Pittman et al. 2003) and use of conversion factors (Bagatto et al. 2002; Scollie et al. 2011) that allow for fewer in situ measurements.

The third procedure is based on the loudness model developed by Moore and Glasberg (1997, 2004). This procedure, termed the Cambridge procedure for loudness equalization, or CAMEQ, has two goals: (1) to give an overall loudness that is similar to or slightly lower than what would be perceived by a normal-hearing person listening unaided and (2) to make all frequency components in speech equally loud, on average, over the range 500–4,000 Hz. The most recent version of this method, CAM2 (Moore et al. 2010), prescribes target gains for a wide frequency range. That feature has been shown to improve speech clarity and recognition of specific high-frequency phonemes compared to narrower bandwidth amplification (Füllgrabe et al. 2010; Moore and Füllgrabe 2010; Moore and Sek 2013). One caveat is that the benefit of high-frequency audibility presumably requires sufficient high-frequency auditory receptors. To date, CAM2 has not been tested for people with severe high-frequency hearing loss.

6.8.1.2 Frequency Lowering

In cases in which high-frequency audibility cannot be achieved by providing gain (because the loss is too severe, the power of the hearing aid amplifier is limited, or acoustic feedback limits the available gain), frequency lowering can be used to shift the frequency of the input signal to a lower frequency region. In a recent survey, a majority of audiologists were reported to use frequency lowering for some of their patients with high-frequency loss (Teie 2012). With regard to speech perception, the rationale is that improved audibility might improve perception of high-frequency phonemes such as fricative consonants spoken by female and child talkers (Pittman et al. 2003). However, frequency lowering (especially strong frequency lowering that affects a wider frequency range) alters the acoustic characteristics of the shifted phoneme. Accordingly, frequency lowering may be beneficial to a listener when audibility outweighs distortion and detrimental when distortion outweighs audibility (Souza et al. 2013).

6.8.1.3 Amplitude Compression

Hearing aids fitted with individual frequency gain shaping have been highly successful at improving speech audibility and perception relative to unaided listening. However, most listeners with hearing loss have threshold elevation without corresponding elevation of their loudness discomfort level. To improve speech perception, the hearing aid must adjust the applied gain depending on the input level of the signal. Accordingly, amplitude compression is a feature of all modern hearing aids. Compression works as follows. The incoming signal is first filtered into a number of frequency bands. The level in each band is estimated. For levels falling below the compression threshold, a fixed (maximum) gain is usually applied (linear amplification). Some very low level sounds (below 30 or 40 dB SPL) may receive less gain (expansion) to reduce the annoyance of environmental sounds or microphone/circuit noise. When the level in a given band exceeds the compression threshold, progressively less gain is applied as the input level increases. The extent to which gain is reduced is determined by the compression ratio.

Compression makes intuitive sense as a means of improving speech perception because higher level inputs require little to no amplification to make them audible. Also, reduced gain for high input levels is needed to avoid loudness discomfort. Regardless of the prescriptive procedure that is used, compression hearing aids are quite successful at achieving improved audibility of low-level sounds and acceptable loudness for high-level sounds (Jenstad et al. 1999, 2000). In a small number of cases (usually severe hearing loss), the auditory threshold is too high—or the loudness discomfort threshold is too low—to achieve audibility across a range of speech levels without using unacceptably high compression ratios. The combined effect of a high compression ratio and fast compression speed may be unacceptable if the resulting processing dramatically alters the intensity relationships between individual sounds and removes the natural intensity contrasts in speech. In those cases, clinical goals often shift to giving good audibility of conversational speech (but not low-level speech) without discomfort from intense sounds.

In a compression hearing aid, the gain changes over time depending on the level of the signal relative to the compression threshold. The speed with which those adjustments occur is determined by the attack and release times of the compressor. While attack times are usually short—typically, 5 ms or less—release times vary widely, from about 10 ms to several seconds. When coupled with a low compression threshold, short attack and release times improve speech audibility by providing more gain for brief, low-intensity speech sounds. However, that improved audibility comes at the expense of altered amplitude properties of the speech signal (Jenstad and Souza 2005). In other words, there may be a trade-off between improved consonant audibility and a desire to retain some natural amplitude variations.

There is unlikely to be a single "best" compression speed that suits all hearing aid wearers. Rather, the optimal compression speed is likely to depend on both the environment and the listener. For example, the detrimental effects of fast compression may be more apparent when there is less acoustic information in the signal, such as for speech that is time compressed (Jenstad and Souza 2007) (mimicking rapidly spoken speech); spectrally degraded (Souza et al. 2012a) (mimicking a listener with poor spectral resolution; see Sect. 6.4.2); or when the listener is more susceptible to signal distortion (see Sect. 6.8.4). Finally, although short release times may offer greater speech recognition benefits for some listeners (e.g., Gatehouse et al. 2006), most listeners prefer a long release time for sound quality (Hansen 2002; Neuman et al. 1998).

6.8.2 Maintaining Acoustic Fidelity

A general assumption has been that once cues are made audible via appropriate frequency-dependent amplitude compression and frequency lowering, they will be accessible to the hearing aid wearer. If audibility were the only requirement for good speech perception, this would be a simple solution. However, amplitude compression and frequency lowering involve distortion of the signal and a loss of fidelity. In some sense, acoustic fidelity can be considered to be traded for audibility. Technically, it would be possible to make every signal audible for every listener but doing so might require amplification parameters (high gain, skewed frequency response, and extreme compression) that would degrade the acoustic signal. Instead, parameters must be chosen to improve the audibility on which speech perception depends while minimizing distortion.

It seems likely that poor spectral resolution for most people with hearing loss will force greater reliance on temporal information (Lindholm et al. 1988; Hedrick and Younger 2007). Each hearing-impaired listener uses both spectral and temporal information, but the balance between the two may vary across listeners. Consider two hypothetical listeners, both with moderately severe sensorineural loss and a 40-dB dynamic range. Listener A has good frequency selectivity and can access a full range of spectral cues to speech, including vowel spectra, formant transitions, and overall spectral shape. Listener B has broadened auditory filters and is limited to coarse representations of spectral information. Listener B must depend to a greater extent on temporal cues, including the amplitude envelope and periodicity in the signal. A clinician might be tempted to adjust hearing aid parameters for both listeners with audibility as the primary goal, using fast-acting wide dynamic range compression (WDRC) to improve the audibility of low-intensity sounds. Although fast-acting WDRC improves audibility, it also distorts the amplitude envelope and may be a poor choice for improving speech perception for Listener B (Jenstad and Souza 2005; Davies-Venn and Souza 2014). Audibility can also be improved by using a higher number of compression channels (Woods et al. 2006), but too many channels will smooth frequency contrasts (Bor et al. 2008) and may be a poor choice for improving speech perception for Listener A. Although such arguments are speculative, a necessary first step in clarifying these issues is to understand how reliance on spectral and temporal properties varies among individuals with hearing loss.

6.8.3 Listening in Noise

Because a common speech perception complaint is difficulty when listening in noise, considerable effort has gone into this aspect of hearing aid design. Two general strategies are used to reduce background noise: directional microphones and digital noise reduction. A more complete discussion of each feature is available elsewhere in this volume (Launer, Zakis, and Moore, Chap. 4; Akeroyd and Whitmer, Chap. 7). Here, the effects of each on speech perception are considered.

6.8.3.1 Directional Microphones

Directional microphones have been used to improve speech perception for nearly 40 years (Sung et al. 1975). Directionality is usually achieved by processing the outputs of two (or three) omnidirectional microphones. This has become a near-universal feature of hearing aids, with the exception of some aid styles (such as completely-in-canal aids) for which directional information is not preserved at the hearing aid microphone(s). A common configuration is a microphone that is both automatic and adaptive, where the modulation pattern and spatial location of the incoming signal are used to activate either an omnidirectional response or a directional response with a specific polar plot (Chung 2004). Because directional microphones operate in the spatial domain, they are successful at improving speech perception when speech and interfering sources are spatially separated. The improvement in SNR can be about 5 dB, which translates to as much as a 30% improvement in speech intelligibility. Directional microphones are less advantageous in cases of multiple or moving noise sources, when the user wishes to switch attention between sources at different azimuths, when the speech signal of interest is behind the user, or in high levels of reverberation (Bentler and Chiou 2006b; McCreery et al. 2012; Ricketts and Picou 2013).

6.8.3.2 Digital Noise Reduction

Digital noise reduction is intended to remove noise while retaining speech information. Digital noise reduction is a nearly universal feature in modern hearing aids, although the type of digital noise reduction and the extent to which it is applied vary markedly. Noise reduction usually involves classifying the signal in each frequency band as predominantly speech or predominantly noise and decreasing the gain in bands that are dominated by noise while preserving the gain in bands that are dominated by speech. Typically, the modulation pattern of the signal is used to estimate whether speech or noise dominates in each band (Bentler and Chiou 2006a; Chung 2012). One limitation is that digital noise reduction cannot function perfectly without a template of the speech alone—something that is not available in real environments. On occasion, digital noise reduction may misclassify within-band noise as speech or misclassify within-band speech as noise. Such processing errors are more likely in cases in which the "noise" comprises other people speaking.

Patient expectations for digital noise reduction are high, but for many years the evidence suggested that it did not improve speech perception (Bentler 2005; Palmer et al. 2006; Bentler et al. 2008). Recently, however, researchers have begun to measure listening effort rather than speech identification. Those studies have consistently found that digital noise reduction reduces listening effort and fatigue and increases acceptance of background noise (Sarampalis et al. 2009; Hornsby 2013; Lowery and Plyler 2013; Gustafson et al. 2014). Because it reduces listening effort, noise reduction may also free cognitive resources for other tasks, such as learning new information (Pittman 2011).

6.8.4 Choosing Hearing Aid Parameters to Suit Individual Cognitive Abilities

Hearing aid choices and parameters have long been customized to suit the patient's pure-tone audiogram and loudness discomfort levels. More recently, it has been recognized that individual cognitive abilities may also be relevant in selecting the parameters of hearing aid processing. Most of that work has relied on measurements of working memory (described in Sect. 6.5). Recall that low working memory is thought to reduce the ability to adapt to a degraded or altered acoustic signal. When hearing aids are used, the signal processing may significantly alter and/or degrade the speech signal. Such signal processing includes WDRC with a short release time (Sect. 6.8.1.3), frequency lowering (Sect. 6.8.1.2), and digital noise reduction (Sect. 6.8.3.2) in cases where classification errors result in reduced fidelity of the target speech signal or where the processing introduces spurious amplitude fluctuations that may affect intelligibility. Lower working memory is associated with poorer performance with short compression release times (Gatehouse et al. 2006; Lunner and Sundewall-Thoren 2007; Souza and Sirow 2014), and higher frequency compression ratios (Arehart et al. 2013a). There is emerging evidence that working memory may affect the benefit of digital noise reduction. One study showed that working memory was modestly associated with speech recognition benefit of digital noise reduction (Arehart et al. 2013b); another showed that digital noise reduction reduced cognitive load but only for listeners with high working memory (Ng et al. 2013). A third study showed no relationship between working memory and speech recognition, but patients with low working memory preferred stronger noise reduction settings (Neher et al. 2014).

Because noise can be a significant problem for patients with lower working memory, it seems probable that, for such patients, the beneficial effects of suppression of noise might outweigh the deleterious effects of the distortion produced by the noise suppression. Because the few data available employed different outcome measures (speech recognition, word recall [i.e., memory load], and overall preference), additional work is needed to clarify the extent of the relationship between working memory and noise reduction. More generally, the relationships between individual cognitive abilities and benefit from different features of hearing aid processing reflect the importance of understanding not only the acoustic effect of the hearing aid but also the interaction of those effects with the listener.

6.9 Summary

For people with normal hearing, speech perception appears largely effortless and occurs unconsciously. Hearing loss can greatly increase the effort involved in understanding speech such that speech perception rises to the level of conscious attention. And, when hearing loss impairs speech perception, it does so in unpredictable ways.

When no hearing aids are used, the consequences of hearing loss vary from minimal effects in selected situations to substantial difficulty such that communication becomes a struggle that impairs every aspect of work and social engagement. Speech perception is determined by both auditory and cognitive factors, ranging from the amount of hearing loss and the specific pattern of auditory damage to the listener's ability to compensate for reduced auditory cues using cognitive processing. Hearing aids can compensate for reduced audibility in many situations but are limited as to how much they can improve communication in adverse conditions, such as for speech in background sounds or reverberation. Although many research studies have defined the general effects of hearing loss (and hearing aids) on speech perception, the variability among individuals serves as a reminder that each individual—and the optimal hearing aid processing for that individual—must also be treated as unique.

Conflict of interest Pamela Souza declares that she has no conflict of interest.

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Chapter 7
Spatial Hearing and Hearing Aids

Michael A. Akeroyd and William M. Whitmer

Abstract The questions of whether hearing-impaired listeners are also impaired for the localization of sounds and what benefits hearing aids can provide are important for understanding the wider effects of hearing impairment. We review here 29 studies published since 1983 that have measured acuity for changes in the horizontal-plane direction of sound sources. Where possible, performance is quantified by the root-mean-square error in degrees. Overall, the results demonstrate that (1) hearing-impaired listeners have poorer left–right discrimination than normal-hearing listeners, by 5° when averaged across all experiments, although there is considerable variation across listeners and experiments; (2) hearing aids lead to a deficit of just 1°; (3) directional microphones relative to omnidirectional microphones give a deficit of 3°; (4) custom form factors have no effect relative to the behind-the-ear style; (5) acclimatization gives a benefit of 1°; (6) a unilateral fitting results in a localization deficit of 5° on average, and the deficit can reach nearly 20°; and (7) hearing-impaired listeners are particularly prone to front–back confusions; hearing aids do nothing to reduce these and sometimes increase them. Although statistically significant effects of hearing aids on localization have been reported, few of them are generalizable, as they often occurred for just some source directions, stimuli, hearing aid features, or groups of listeners. Overall, there is no experimental evidence for a benefit from hearing aids for directional acuity.

Keywords Acclimatization • Bilateral aiding • Behind-the-ear • Directional microphones • Front–back confusions • Hearing aids • Localization • Meta-analysis • Minimal audible angle • Unilateral aiding

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7.1 The Fundamental Phenomena of Spatial Hearing

The direction of a sound source, be it to the left or right, front or back, or above or below the listener, is derived mainly by integrating information from the two ears. The auditory system uses two primary cues—interaural time differences (ITDs) and interaural level differences (ILDs)—for determining most directions (e.g., Durlach and Colburn 1978; Blauert 1997). The difference in path distance between the two ears for a sound directly to the right of the listener is about 23 cm for adults. Because sound takes about 30 μs to travel 1 cm, the sound reaches the left ear about 700 μs after it reaches the right ear. This is an ITD. If instead the source is directly to the left of the listener, the sound reaches the right ear about 700 μs after it reaches the left ear; intermediate directions result in intermediate ITDs. The ITD is nonzero for any direction that is not on the exact midline between the ears (i.e., not ahead, above, below, or behind). For a pure tone, an ITD leads to an interaural phase difference (IPD): the IPD in degrees is equal to the ITD in seconds multiplied by 360 times the frequency of the tone.

For most purposes, a simple equation describes how the ITD in the horizontal plane depends on left–right direction: $\text{ITD} = (R\theta + R\sin\theta)/c$, where θ is the azimuth in radians, R is the radius of the head in meters, and c is the speed of sound in meters per second (Woodworth 1938; Moore 2013). Conventionally, an azimuth of 0 is directly ahead of the listener, a positive azimuth is to the right, and a negative azimuth is to the left. This equation is derived from simple geometry and assumes that the head is spherical, the ears are diametrically opposite each other, and the wavelength of the sound is much smaller than the size of the head and ears (i.e., the frequency of the sound is very high; Kuhn 1977). The equation matches experimental measurements taken on real people using clicks as the sound sources (Feddersen et al. 1957) or on manikins with high-frequency pure tones as the sounds (Kuhn 1987). However, although useful, Woodworth's equation is not accurate in all circumstances. First, the equation fails for low-frequency pure tones, for which the ITDs are larger than predicted (Kuhn 1977; Aaronson and Hartmann 2014). Second, the torso and clothing can affect the ITD (Kuhn 1977; Treeby et al. 2007). Third, the size of the head (R) varies across individuals: for adults, R on average is larger for taller than for shorter people by about 1/6 mm per centimeter of height, is larger for men than for women by about 0.25 cm (Bushby et al. 1992), and is about double the value at birth. Finally, when pondering the effects of evolution in determining the current characteristics of spatial hearing, it is important to bear in mind that the size of the head has also increased substantially through hominid evolutionary history (Tattersall 2008).

Figure 7.1a shows ITDs measured on a manikin, taken from Kuhn (1977) for a clothed torso. The solid line shows data for an azimuth of 15°; the other lines are linearly interpolated values for smaller angles. The asterisks plot the values calculated from $(R\theta + R\sin\theta)/c$, assuming $R = 0.09$ m. It can be seen that the predictions closely match the measured ITDs at high frequencies, but at the lowest frequencies, the measured ITDs are about 33 % larger than predicted. Figure 7.1b shows the inverse relationship, that is, the azimuth needed to give ITDs of 10, 25, 50, and 100 μs. In this range, the azimuth in degrees is roughly equal to one-tenth of the ITD in

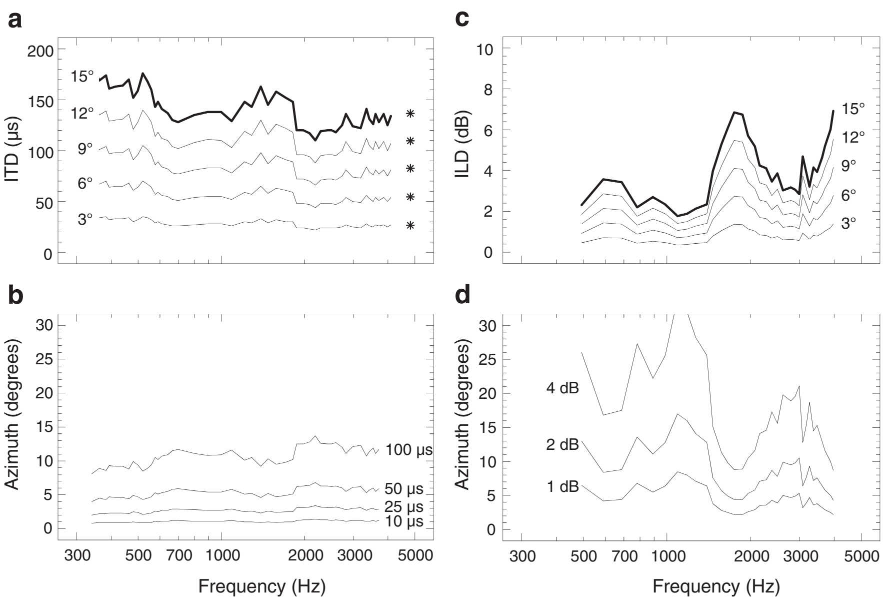

Fig. 7.1 (**a**) The dependence of interaural time difference (ITD) on frequency, for five azimuths close to directly ahead. The *bold line* is taken from Kuhn (1977, Figure 11) for an azimuth of 15°; the other lines are linear interpolations for 3°, 6°, 9°, and 12°. The ITD at 0° is assumed to be zero at all frequencies. The ITDs were taken from measurements of a clothed manikin with microphones at its surface. The *asterisks* plot the predicted values calculated from ITD = $(R\theta + R\sin\theta)/c$. (**b**) The same data but plotted as the dependence of azimuth on frequency for four values of ITD. (**c**, **d**) The corresponding data from Kuhn for ILDs at the same angles

microseconds, so an angle of 1°, which is the approximate value of the smallest detectable change in direction, corresponds to an ITD of about 10 μs.

Because the head casts an acoustic shadow, the level of a sound is generally less at the shadowed ear than at the ear facing the source. The value of the ILD varies with direction and frequency but in a complex manner, and there is no simple formula that describes this. To a first approximation, ILDs are larger at higher than at lower frequencies and are larger for azimuths close to ±90° than azimuths near the midline. There can be sharp dips in ILD at some frequencies but strong peaks at other frequencies, often with a range of about 20 dB. The ILDs for two neighboring directions may bear little resemblance to one another. Figure 7.1c, d shows the interpolated small-angle ILD versus direction relationship for a clothed torso (Kuhn 1977). The large idiosyncratic variation of ILD with frequency is clear. At low frequencies, the azimuth in degrees is approximately five to eight times the ILD in decibels; at high frequencies, the azimuth is about three to five times the ILD.

Values of ILDs are generally obtained from measurements of the head-related transfer functions (HRTFs) on humans or manikins (e.g., Gardner and Martin 1995; Blauert et al. 1998). Because these measurements are commonly performed in an anechoic chamber, they represent the ILDs that would occur only if the listener was in free space with unoccluded pinnae and a source that was not too close. ILDs can be calculated analytically if simplifying assumptions are made, such as the head being an exact sphere (e.g., Rayleigh 1894; Macaulay et al. 2010), or they can be computed from boundary-element models (e.g., Kreuzer et al. 2009). The complexity of ILDs occurs because the level of the sound at each ear depends critically on how sound travels to the facing ear and shadowed ear, which in turn depends on the amount of diffraction round the head and pinna and the interference with any reflections from the shoulders and pinnae. The magnitudes of many of these effects are dependent on the direction of the source and the ratio of the wavelength of the sound to the size of whatever object is causing the diffraction or reflection. Any object that introduces extra reflections affects the ILDs. It would therefore be expected that sitting in a high-back chair, lying on a sofa, putting one's head on a cushion or pillow, being close to a wall, wearing a substantial hat, or putting one's hands on one's head all will generate different ILDs than those that occur when standing in free space away from any reflective surface.

The acuity for detecting changes in ITD or ILD is measured experimentally by the just noticeable difference (JND), which is the smallest change that can be reliably detected. Most commonly the change is relative to a reference ITD or ILD of zero. There is a vast literature on the JND for ITD (JND_{ITD}); Section 7.2 summarizes the main results for hearing-impaired listeners. The key results for normal-hearing listeners are (1) the JND_{ITD} reduces as the sound's duration increases, reaching an asymptote at a duration of about 300 ms (e.g., Tobias and Zerlin 1959); (2) for a wideband noise burst lasting longer than about 300 ms, the JND_{ITD} can be as small as 10 μs (e.g., Klump and Eady 1956); (3) for pure-tone stimuli, the JND_{ITD} depends on frequency, being about 60 μs at 250 Hz, 10 μs at 1,000 Hz, and 20 μs at 1,250 Hz. When the frequency is increased above 1,250 Hz, the JND_{ITD} increases dramatically, becoming unmeasurably large above about 1,500 Hz (e.g., Klump and Eady 1956; Brughera et al. 2013); (4) for complex sounds with substantial modulations in their envelopes, a JND_{ITD} can be measured when all components have high frequencies, and, with the right type of modulation, the high-frequency JND_{ITD} can be as small as the low-frequency JND_{ITD} (e.g., Bernstein and Trahiotis 2002); (5) for a signal in noise, the JND_{ITD} increases as the signal-to-noise ratio decreases (e.g., Stern et al. 1983); and (6) for sounds whose ITD varies across their duration, more weight is placed on ITDs at the start of the sounds than at later times (e.g., Hafter and Dye 1983; Akeroyd and Bernstein 2001).

There are fewer experiments measuring the JND for ILD (JND_{ILD}) for normal-hearing listeners. It is on the order of 1 dB and there is only a small frequency dependence, with a slight worsening around 1 kHz (e.g., Grantham 1984; Hartmann and Constan 2002).

The smallest detectable change in direction is traditionally called the minimum audible angle (MAA; Mills 1958), although here we refer to it as the JND_{MAA} to emphasize that it is a JND. For normal-hearing listeners, the JND_{MAA} can be as low as 1° (see Akeroyd 2014 for some speculations about why the JND_{MAA} is so small). This value is found for changes in azimuth away from 0° using pure-tone stimuli with a frequency close to 750 Hz (Mills 1958). The JND_{MAA} is larger at lower frequencies and medium frequencies, reaching about 3° for frequencies close to 2,000 Hz, and then decreases for frequencies up to about 8 kHz. It is much higher for changes in azimuth around a reference direction close to 90° or −90° (e.g., Mills 1958) or for changes in elevation (e.g., Grantham et al. 2003). The data on the JND_{MAA} for hearing-impaired listeners are considered in Sects. 7.4.2 and 7.4.8.

In principle, it should be possible to use the relationships shown in Fig. 7.1 to predict the JND_{MAA} from the JND_{ITD} or JND_{ILD}, although for any dynamic broadband sound, the calculations will be complicated by the need to know how the auditory system weights and integrates the information across ITD/ILD, frequency, and time. There is strong evidence for such integration, as the auditory system uses both ITDs and ILDs to determine the direction of sound sources in the real world. Certainly there is a nonlinear weighting or integration across time: the various aspects of the precedence effect, such as localization dominance—if two clicks are played in close succession from two different directions, the two are heard as a single sound whose perceived direction is that of the first click alone—demonstrate that for localization the auditory system strongly emphasizes the information at the beginning of a sound (e.g., Litovsky et al. 1999; Akeroyd and Guy 2011). For sounds such as speech that continually change in level and spectrum, ITD and ILD cues occur across many frequencies. It makes sense for the auditory system to use the best information available to determine the direction of the source, as often one frequency region will be partially masked by a background sound, and the most informative frequency region will change from moment to moment.

Neither ITDs nor ILDs wholly determine *three*-dimensional direction. The ITD/ILD of a sound from a source with an azimuth of 10° will be almost the same regardless of whether the source is in front of, above, below, or behind the listener. The locus of all source positions leading to the same ITD or ILD is known as the "cone of confusion" for that ITD or ILD (e.g., Shinn-Cunningham et al. 2000; Moore 2013), and it occurs because the head is essentially front–back symmetric, being approximately spherical with the ears placed at (almost) a diameter [note that Woodworth's $(R\theta + R \sin\theta)/c$ formula is exactly front–back symmetric]. Many experiments have demonstrated that listeners sometimes report a sound from a source that is physically in front as being from behind or vice versa. Hearing-impaired listeners, especially if aided, are particularly susceptible (see Sect. 7.4.9).

Whether a source is actually located in front, behind, up, or down can be determined in two ways. The first is that if the sound has high-frequency components, then the ILDs can be used, as the wavelengths of high-frequency sounds are short enough to interact with the pinnae, which are very much front/back, up/down asymmetric. The second way, which works at all frequencies, is to rotate the head and assess the way that the ITD and ILD change (e.g., Wallach 1940; Brimijoin and Akeroyd 2012). As people are almost always moving their head—for instance, while walking, turning to people when talking in a group, or even just fidgeting—the ITDs and ILDs are almost always changing. It thus seems sensible for the auditory system to continually use motion information to help locate sounds. For example, if a sound source is at 0° azimuth, and the head is rotated 10° to the left, then there will be an ITD that is leading in the right ear by 100–150 μs. But if the sound were really behind, at 180° azimuth, then the same head movement would result in an ITD that was leading in the *left* ear. A comparison of the direction of head movement with the direction of the resulting ITD (or ILD, the same reasoning applies) therefore resolves the problem.

7.2 JNDs for ITD and ILD for Hearing-Impaired Listeners

As the data that follow concentrate on acuity for changes in spatial direction, the experimental results on sensitivity to changes in the underlying cues of ITD or ILD are summarized only briefly. It is generally found that the JND_{ITD} is higher for hearing-impaired listeners than for normal-hearing listeners (e.g., Hawkins and Wightman 1980), although in some studies, the effect may be attributed to age rather than hearing loss (e.g., Strelcyk and Dau 2009). The highest frequency at which listeners can detect a change in the ITD of a pure tone is much reduced: for instance, Neher et al. (2011) reported a mean value of 850 Hz for a group of hearing-impaired listeners ($n=23$) and 1,230 Hz for a group of normal-hearing listeners ($n=8$). This effect is of importance to the integration of ITD cues across frequency, as it implies that the upper frequency limit is considerably lower for hearing-impaired listeners than for normal-hearing listeners.

There is evidence that age and hearing impairment affect interaural sensitivity independently. For 46 listeners with hearing impairment, King et al. (2014) reported a significant correlation of 0.42 between thresholds for detecting changes in IPD at 500 Hz and hearing thresholds at 250 and 500 Hz, after partialing out the effect of age. Hawkins and Wightman (1980) reported mean JND_{ITD}s that were nearly three times higher for a small group of hearing-impaired listeners ($n=7$) than for a small group of normal-hearing listeners ($n=3$) of essentially the same mean age. King et al. (2014) reported a correlation of 0.45 between age and JND_{ITD}s at 500 Hz after controlling for hearing loss, and Gallun et al. (2014) found a statistically significant difference in ITD sensitivity between a younger, normal-hearing group (mean age 29 years) and an older group with normal-to-minimal hearing losses (mean age 59 years). Ross et al. (2007) found a substantial effect of age on the highest frequency at which ITD changes could be detected: the mean values for groups of young adults ($n=12$), middle-aged adults ($n=11$), and older adults ($n=10$) were 1,203, 705, and 638 Hz, respectively. The mean ages of their groups were 27, 51, and 71 years, respectively (the mean hearing losses, for frequencies up to 2 kHz, were 8, 13, and 19 dB, respectively).

Results obtained with the temporal fine-structure low-frequency (TFS-LF) test (Hopkins and Moore 2010, 2011; Moore 2014) are also relevant as this test measures the JND_{IPD} for a pure tone, which is linearly related to the JND_{ITD}. To take just two recent examples, Füllgrabe et al. (2015) compared TFS-LF performance for older ($n=21$; mean age$=$67 after) and younger adults ($n=9$; mean age$=$23 years). Importantly, the two groups had audiometrically normal hearing, always less than or equal to 20 dB hearing level (HL), and mostly less than or equal to 10 dB. Performance for the older group, measured by the detectability index, d', was about half that for the younger group. The correlation with HL was insignificant. A second example is from Whitmer et al. (2014), who reported a correlation of 0.60 between TFS-LF test performance and age, after partialing out the effect of hearing loss, for a group of listeners aged 26 to 81 years ($n=35$; hearing losses $=-1$ to 67 dB HL). The corresponding correlation between TFS-LF performance and hearing loss, after partialing out the effect of age, was 0.25. These results further indicate that age and hearing loss affect acuity for changes in interaural cues.

There are very few experiments measuring the JND_{ILD} for hearing-impaired listeners. One example, with just four listeners, reported JND_{ILD}s of 2–8 dB, which are larger than the typical value of 1 dB for normal-hearing listeners (Gabriel et al. 1992).

7.3 Experimental Measures of Directional Acuity

There are two main methods by which the acuity for spatial direction can be assessed. In one, "source discrimination," two sounds (typically) are presented from loudspeakers with slightly different directions, ϕ_1 and ϕ_2. The listener's task is to decide if the order of presentation was ϕ_1 then ϕ_2 or ϕ_2 then ϕ_1. The difference between ϕ_1 and ϕ_2 varies across trials, either following a randomized, predefined list, giving a psychometric function from which a threshold can be calculated, or following an adaptive track, giving a threshold directly. The threshold is the JND_{MAA}. Somewhat surprisingly, only three experiments have measured JND_{MAA}s for hearing-impaired listeners (Häusler et al. 1983; van Esch et al. 2013; Brimijoin and Akeroyd 2014). These are difficult experiments to perform; the JND_{MAA} can be as low as 1°, so, if real sound sources are used, one needs either a set of loudspeakers spaced at least that close or a boom that can silently move a single loudspeaker accurately and with high resolution in complete darkness to avoid the listener seeing the boom move. It is also necessary to measure accurately where the listener's head actually is. Each 1 cm of inaccuracy in the position of a listener's head gives an inaccuracy of 0.6° in azimuth for a loudspeaker placed 1 m away, which is not an uncommon experimental distance.

An alternative is to use the techniques of virtual acoustics to synthesize direction with head-related transfer functions (HRTFs). This can simulate presentation of sounds from any direction, especially if interpolation is used to create HRTFs for directions intermediate to those measured. However, care needs to be taken experimentally for this to work properly and to be trusted to give accurate directional information. The concern is that the details of the HRTF, especially the high-frequency ILDs, depend on how the sounds diffract and interfere with the pinnae, and pinnae differ from individual to individual. If the HRTFs are obtained from a set of pinnae, such as those on a manikin or those of a different person, that differ too much from the listener's own pinnae, then the direction perceived by the listener may not exactly match the direction used to record the HRTF, especially in elevation. It is better to use HRTFs measured for each listener, although making all the individual measurements is experimentally intensive and so the alternative practice of using premade manikin or other-listener recordings (e.g., Gardner and Martin 1995; Blauert et al. 1998) is quite common. Only three spatial-hearing experiments on hearing impairment used virtual acoustics, and all three used nonindividualized HRTFs (Chung et al. 2008; van Esch et al. 2013; Brimijoin and Akeroyd 2014).

In the other method, "source identification," a sound is presented from any one of a set of loudspeakers (Fig. 7.2). For maximum accuracy, one needs loudspeakers whose spacing leads to graded performance between chance and perfect (separations of 10°–20° are typical, though it is questionable if these are fine enough). Source identification has been a relatively popular method but with many variations in experimental details; for instance, the loudspeakers can be visible (e.g., Lorenzi et al. 1999b; Drennan et al. 2005) or concealed behind a curtain or gauze (e.g., Keidser et al. 2006; Freigang et al. 2014); the response can be reporting the direction of the relevant loudspeaker (e.g., Neher et al. 2011) or giving the direction per se by pointing the head toward the sound (e.g., Best et al. 2011), or by using a flashlight (Freigang et al. 2014). Given the modern improvements in motion-tracking

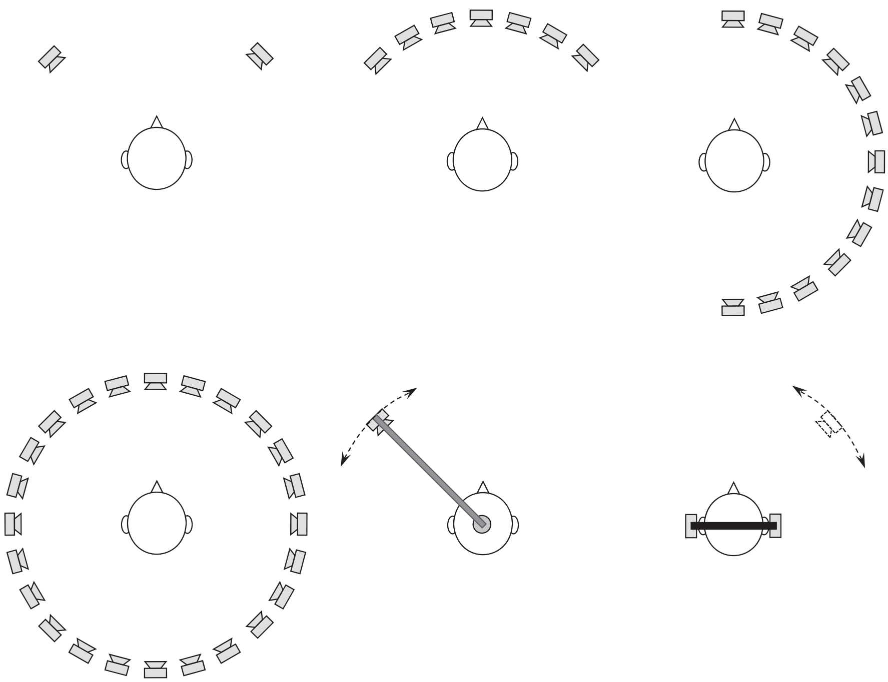

Fig. 7.2 Schematic illustrations of example loudspeaker arrays that have been used in experiments on sound localization. The *top row* shows a symmetric pair, an arc, and a semicircle; the *bottom row* shows a full circle, a single loudspeaker on a moveable boom, and virtual acoustics using headphones

technology along with its ever-cheaper cost (for instance, it can be done using the remote control of a Nintendo Wii videogame console; see Brimijoin et al. 2013), pointing responses are recommended for future experiments (cf. Brungart et al. 2000), but care must be taken to ensure that the response starts after the sound stops, as otherwise the motion of the listener's head creates changes in the ITDs and ILDs.

The source discrimination and source identification methods differ in that the former requires a relative judgment (is the first sound to the left or right of the second sound?), whereas the second requires an absolute judgment (what is the direction of the sound?). It would be expected that they would give results that were strongly related, as both are taken as measures of the fundamental acuity of directional hearing. Only a very few experiments have explicitly made the comparison of one against the other, and those that have done this have given diverging results, for instance, Recanzone et al. (1998) reported correlations of 0.5–0.8 between scores for the two methods, while Freigang et al. (2014) reported correlations of −0.15 to 0.43 across a set of experimental conditions. One potential issue is that many reported errors were smaller than the angular spacing between loudspeakers in an array. More research is needed to clarify the relationship between the two methods.

In the source identification method, the loudspeaker used to present the sound varies from trial to trial. If N trials are presented from the loudspeaker at azimuth L and on trial i the listener responds with the azimuth A_i, then the error e_i on each trial is

$$e_i = A_i - L$$

where all values are in degrees. Various descriptive statistics have been used to summarize performance. They include the fraction of trials (F) for which the response is correct (i.e., the subset for which $e_i = 0$); the average across all N trials of the mean error, E; the mean absolute error, E'; and the root-mean-square (RMS) error, D (the letters E and D were introduced by Hartmann 1983). These measures are defined mathematically as

$$E = \sum_{N}^{i=1} \frac{e_i}{N}$$

$$E' = \sum_{N}^{i=1} \frac{|e_i|}{N}$$

$$D = \sqrt{\sum_{N}^{i=1} \frac{e_i^2}{N}}$$

For the across-experiment averages reported below, we used D weighted by the number of listeners. The averages are termed D_N. This was done to compensate for the large differences in group sizes, as it seems unfair to weight a mean result based on (in one example) 4 listeners equally with another mean result based on 72 listeners.

It is recommended that D be reported in all studies of localization performance (Hartmann 1983; Yost et al. 2013). It has been the most popular of the various possible measures, has units of degrees so is immediately interpretable, is equal to 0 for perfect performance, is affected by both random direction errors and overall systematic biases, and is similar to the standard deviation. However, specific experimental situations sometimes require the use of some other measure.

Three of the measures are related to each other (Fig. 7.3). Each panel plots a subset of the data collected by Akeroyd and Guy (2011) for 93 listeners with hearing losses ranging from none (0 dB HL) to moderate (60 dB HL). The data presented here are for a control condition in which one word was presented in each

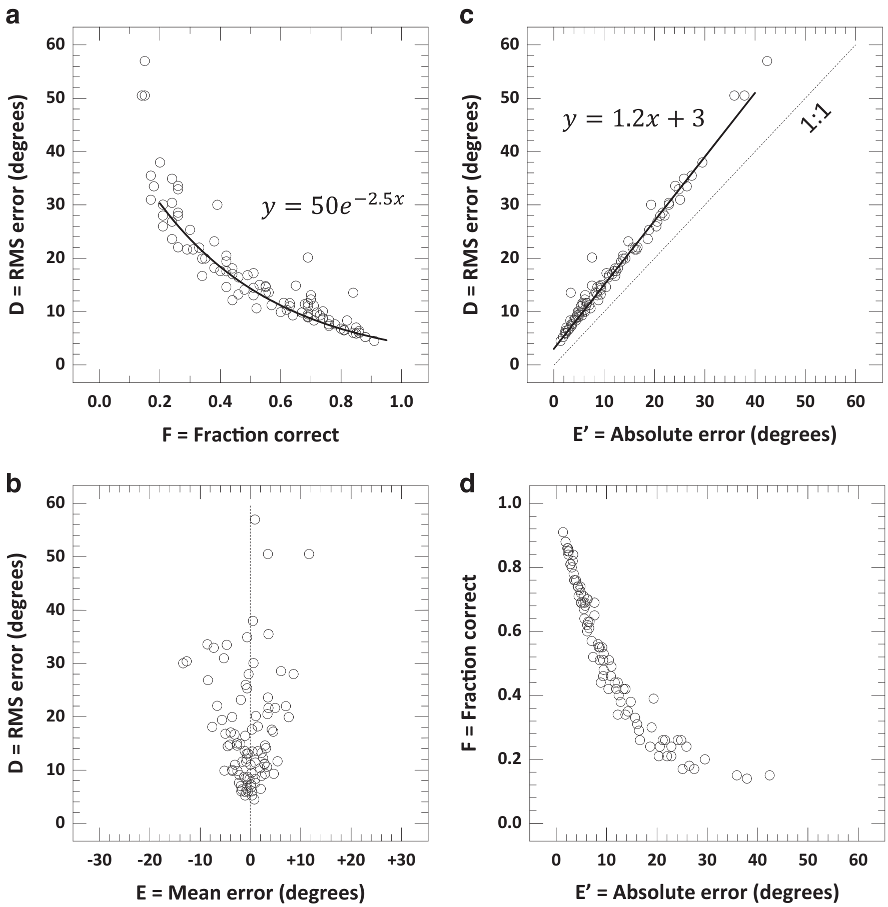

Fig. 7.3 (**a**, **b**, **c**) The relationship between RMS error D and the fraction correct F, mean error E, and absolute error E'. The fits in the top panels are based on simple formulae and rounded parameters. (**d**) The relationship between fraction correct F and absolute error E'. The data are taken from Akeroyd and Guy (2011) for 93 listeners

trial. The source identification method was used, with 12 presentations each from 13 directions from −90° to +90° azimuth at 15° intervals. It can be seen from Fig. 7.3 that the RMS error *D* is essentially a linear function of absolute error *E*′ (Fig. 7.3c), whereas it is a curvilinear function of fraction correct *F* (Fig. 7.3a) and is essentially unrelated to mean error (Fig. 7.3b). The latter is to be expected, as the squaring operation in the RMS calculation means that either positive or negative mean errors can give the same value of *D*.

The solid lines in Fig. 7.3a, c are empirical fits to the data based on simple formulas and with parameters rounded from least-square best fits

$$D = 50e^{-2.5F}$$

$$D = 1.2E' + 3$$

where *F* is a fraction and both *D* and *E*′ are in degrees (see also Fisher 1922; Geary 1935). These equations were used in the analyses that follow to transform published data from those measures to *D*.

Noble and Byrne's set of experiments used an idiosyncratic statistic, "error score." This is a form of absolute error, calculated as the difference, in number of loudspeakers, between the actual source direction and the response direction. It sometimes included the product of horizontal and vertical errors (e.g., Noble and Byrne 1990, 1991), so confounding performance along those planes and the definition changed in later years (Byrne et al. 1998).

Another method of presenting the data is to show scatterplots of response direction versus target direction, sometimes with the size of the symbol proportional to the number of responses (e.g., van den Bogaert et al. 2006; Best et al. 2010). This method shows clearly any response biases or front–back confusions, but it has the disadvantage that deriving quantitative measures for meta-analysis is particularly difficult.

When considering directional acuity in the front–back dimension, it is important both to analyze responses in terms of RMS error *D* and to consider front–back confusions. Note that the term "front–back" covers both front-to-back and back-to-front confusions; the former are more common than the latter (e.g., Best et al. 2010; van den Bogaert et al. 2011).

Two other points deserve notice. First, the group sizes vary considerably across the studies considered here, from 4 (Lorenzi et al. 1999b) to 72 (van Esch et al. 2013). Yost et al. (2013) noted that 80 % power for detecting a difference in *D* of 1.5° at $\alpha = 0.05$ requires 12 listeners (see also Jiang and Oleson 2011). Of the experiments considered here, only Jensen et al. (2013) and Kuk et al. (2013) reported a priori power analysis. Given that statistical significance is affected by the number of listeners, the analyses that follow thus emphasize effect sizes. A sense of scale of can be gained from knowing that one's index fingernail, held at arm's length, subtends an angle of about 1° and one's hand subtends about 10°. It seems reasonable to expect that a difference in localization error on the order of 1° will not be relevant in everyday life, but a difference of 10° likely will be. Moreover, it is arguable that any small differences, up to about ±2°, are within the range of measurement error.

Also, many of the data summarized below have been taken from published graphs and this introduces errors of 1–2°. The errors for the data converted to *D* are probably larger still given that they are based on an empirical fraction-correct/absolute-error transform.

Second, the mean results of a group may not apply to every member of that group. Even when the mean performance of a hearing-impaired group is worse than that of a normal-hearing group, it is generally not the case that every hearing-impaired *individual* performs more poorly. An example is shown in Fig. 7.4a. This shows the distribution of JND_{MAA} values from Häusler et al. (1983) for a group of normal-hearing listeners (plotted above the horizontal line) and a group of hearing-impaired listeners with various kinds of sensorineural hearing loss (plotted below the line). The data are thresholds for detecting changes from 0° azimuth. It can be seen that the distribution for the hearing-impaired listeners is at least twice as broad as that for the

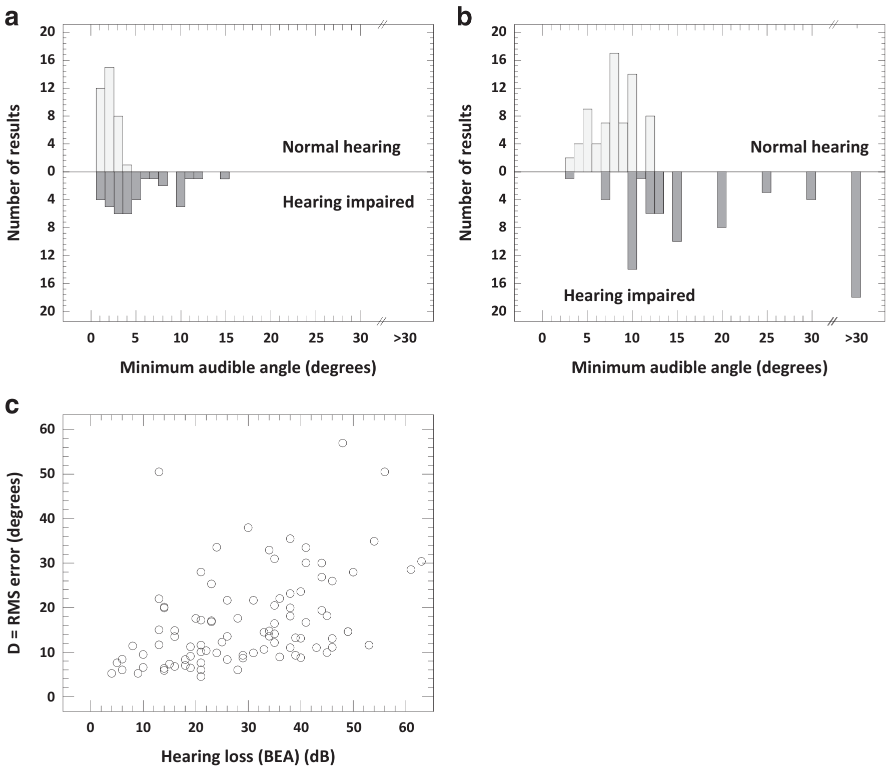

Fig. 7.4 (**a**) The distribution of individual left–right JND_{MAA} values for normal-hearing and hearing-impaired listeners (above and below the *horizontal line*, respectively). The data are from Häusler et al. (1983). (**b**) Distribution of individual front–back JND_{MAA} values from the same study. (**c**) Left–right RMS error for words in babble as a function of better-ear average (BEA) hearing loss. The data are taken from Akeroyd and Guy (2011)

normal-hearing listeners. Moreover, the two distributions overlap considerably. Figure 7.4b shows the corresponding distributions for the detection of changes from +90° azimuth. Although the overlap is less, it is still present and the range of scores is far wider. Figure 7.4c shows the values of *D* from Akeroyd and Guy (2011), plotted as a function of each individual's average hearing loss across 500, 1,000, 2,000, and 4,000 Hz in the better ear (termed the better-ear average [BEA]). There is a moderate correlation of 0.44 between *D* and the BEA, but the variation at each level of hearing loss is substantial, and some people with losses up to 50 dB HL can give reasonably low values of *D*.

7.4 Spatial Acuity for Unaided and Aided Hearing-Impaired Listeners

This section collates the data from 29 experiments that have measured the acuity of directional perception in the horizontal plane for hearing-impaired listeners with bilateral, symmetric (or mostly so) sensorineural hearing losses. Figure 7.5 schematically illustrates the geometry; Table 7.1 summarizes which studies contributed data to the main analyses. Sections 7.4.2–7.4.7 cover results concerned with the acuity for deciding whether one sound is to the left or right of another. The sounds themselves were presented from in front (e.g., Best et al. 2011), behind (e.g., Vaillancourt et al. 2011), or to one side (e.g., Neher et al. 2011). The JND_{MAA} data for sounds presented from ahead are included in Sect. 7.4.2. Sections 7.4.8 and 7.4.9 cover the results for deciding if a sound is in front or behind, including data on sounds presented from the side and analyzed in terms of RMS error along the front–back dimension, as well as results on the JND_{MAA} for sounds presented from the side.

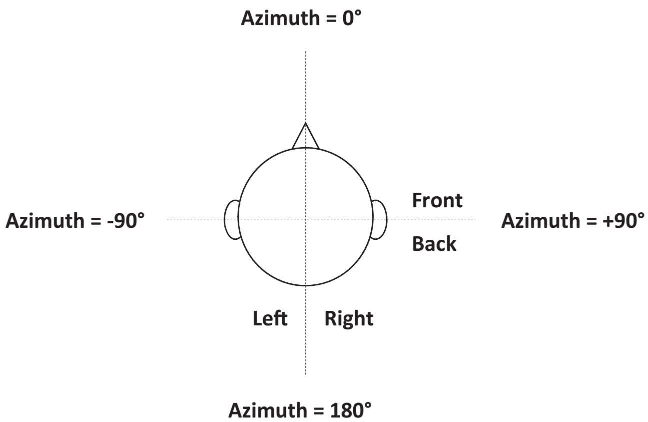

Fig. 7.5 The definition of left, right, front, back, and azimuth

Table 7.1 Key elements of the experimental design of the 29 studies included in the surveys

Study	Number of loudspeakers	Arrangement	Loudspeaker separation	Measure
Abel and Hay (1996)	6	Full circle	60°	%
Akeroyd and Guy (2011)	13	Full circle	15°	RMS
Best et al. (2010)	1	Boom	NA	AE
Best et al. (2011)	11	Arc	10°	RMS
Brungart et al. (2014)	26	Three arcs	38°	AE
Byrne et al. (1992)	11	Semicircle	15°	ES
Byrne et al. (1996)	11	Semicircle	15°	ES
Byrne et al. (1998)	11	Semicircle	15°	ES
Chung et al. (2008)	VA	Headphones	22.5°	AE
Drennan et al. (2005)	11	Semicircle	18°	RMS
Häusler et al. (1983)	1	Boom	NA	MAA
Jensen et al. (2013)	13	Semicircle	15°	RMS
Keidser et al. (2006)	21	Full circle	18°	RMS
Keidser et al. (2007)	21	Full circle	18°	RMS
Keidser et al. (2009)	21	Full circle	18°	RMS
Köbler and Rosenhall (2002)	8	Full circle	45°	%
Kuk et al. (2014)	12	Full circle	30°	%
Lorenzi et al. (1999b)	11	Semicircle	18°	RMS
Neher et al. (2011)	13	Semicircle	15°	RMS
Noble et al. (1994)	11	Semicircle	15°	%
Noble et al. (1997)	11	Semicircle	15°	ES
Noble et al. (1998)	11	Semicircle	15°	%
Picou et al. (2014)	4	Arc	15-90°	%
Simon (2005)	12	Arc	3°	AE
Smith-Olinde et al. (1998)	VA	Headphones	22.5°	RMS
Vaillancourt et al. (2011)	12	Semicircle	15°	AE
van den Bogaert et al. (2006)	13	Semicircle	15°	RMS
van den Bogaert et al. (2011)	13	Semicircle	15°	RMS
van Esch et al. (2013)	VA	Headphones	NR	MAA

AE, absolute error, ES, error score, MAA, minimum audible angle, NA, not applicable, RMS, root-mean-square, VA, virtual acoustics; %, percentage (or fraction) correct

Other major reviews of the effects of hearing impairment and hearing aids on directional hearing include Byrne and Noble (1998) and Durlach et al. (1981). For data on listeners with unilateral losses, conductive losses, or normal-hearing listeners with earplugs or hearing aids, see Noble et al. (1994) and Wiggins and Seeber (2011). All the experiments reported here consider the direction of a source: summaries of the few experiments that have considered the apparent width of a source are in Whitmer et al. (2012, 2014). Also, the data are limited to the detection of changes in the horizontal plane at zero or small elevations.

7.4.1 The Expected Effects of Hearing Aids

Hearing aids could potentially improve sound localization acuity by increasing audibility, and directional microphones might help to resolve front–back confusions, as they generally reduce the level of sounds from behind; see Keidser et al. (2006).

Hearing aids might also degrade sound localization. First, whenever the hearing aid on the left ear has different characteristics than the hearing aid on the right, such as when the aids are in different processing modes (see Launer, Zakis, and Moore, Chap. 4) or have different amounts of gain due to the independent operation of multichannel compression at the two ears (see Chap. 4 and Souza, Chap. 6), spurious interaural differences will be introduced. For detailed measurements of the effects of hearing aids on ITDs and ILDs, see Keidser et al. (2006) and van den Bogaert et al. (2006). A unilateral fitting is an extreme example of this. Second, anything that distorts or modifies spectral pinna cues may adversely affect directional perception. For example, the microphones of a behind-the-ear (BTE) hearing aid are behind the pinna so the pattern of sound diffraction and interference around the pinna is different from "normal," whereas an in-the-ear (ITE) hearing aid with a microphone at the entrance to the ear canal provides pinna cues that more closely match the normal cues. For studies of the effects of different hearing aid styles (see Moore and Popelka, Chap. 1), see Jensen et al. (2013) and van den Bogaert et al. (2011). Furthermore, if there are any effects of acclimatization, then data collected immediately after the initial fitting may differ considerably from those collected after a few weeks.

Unless the leakage of sound past the earmold is very small, the ITDs and ILDs that the listener experiences result from mixing of the aided and unaided (leakage) sounds (see Chap. 1). The aided signal is delayed by 1–10 ms (see Chaps. 1 and 4, and Zakis, Chap. 8). The mixed signal will have a spectral ripple that may not be matched at the two ears, leading to spurious ILDs. There is also the question of whether localization based on ITDs would be affected by the precedence effect (Litovsky et al. 1999; Akeroyd and Guy 2011). Given that the delayed sound from the aid will be more intense than the leakage sound, this seems unlikely, although it might occur for frequencies at which the aid produces little gain. Experimental tests of the precedence effect with hearing aids are needed but are currently lacking, excepting an abstract by Seeber et al. (2008) on linear and compression hearing aids.

7.4.2 Hearing Loss and Left–Right Acuity

Twelve studies have compared the detection of changes in azimuth for mostly older listeners with roughly symmetrical bilateral sensorineural hearing loss (SNHL) and mostly younger normal-hearing listeners (see Table 7.2). Figure 7.6a summarizes the data as a scatterplot of the values of *D* for the impaired group (*y*-axis) versus *D* for the normal group (*x*-axis), with one point per separately reported experimental result. Some studies ran multiple stimulus conditions so there are more than 12 points. The three diagonal lines indicate performance for the impaired/older group

Table 7.2 Key characteristics of the hearing-impaired listener groups tested by the included studies and the analyses in which the data are included

Study	Number of HI listeners	Age range, years	Mean or median age, years	Present analyses
Abel and Hay (1996)	23	42–73	NR	Fig. 7.6a
Akeroyd and Guy (2011)	93	40–78	63	Figs. 7.4, 7.5c
Best et al. (2010)	11	48–75	60	Figs. 7.6a, 7.8, 7.9b, 7.10a, 7.12a
Best et al. (2011)	7	22–69	NR	Fig. 7.6a
Brungart et al. (2014)	20	27–80	65	Figs. 7.6a, 7.8, 7.9a
Byrne et al. (1992)	≤20	NR	66	Figs. 7.9a, 7.9b, 7.11b
Byrne et al. (1996)	≤12	NR	NR	Figs. 7.8, 7.11a, b
Byrne et al. (1998)	≤22	15–74	53	Figs. 7.8, 7.11a, b
Chung et al. (2008)	8	62–80	66	Fig. 7.10b
Drennan et al. (2005)	7	63–74	68	Figs. 7.8, 7.10a
Häusler et al. (1983)	14	22–79	58	Figs. 7.5a, b, 7.6b
Jensen et al. (2013)	17	42–73	63	Figs. 7.9b, 7.11b
Keidser et al. (2006)	12	37–78	75	Figs. 7.10a, b, 7.11b
Keidser et al. (2007)	≤14	39–83	71	Fig. 7.9a
Keidser et al. (2009)	21	57–82	76	Figs. 7.6a, 7.8, 7.10a, b, 7.11a, b
Köbler and Rosenhall (2002)	19	64–73	69	Fig. 7.8
Kuk et al. (2014)	9	65–81	73	Fig. 7.8
Lorenzi et al. (1999b)	4	52–74	65	Fig. 7.6a
Neher et al. (2011)	23	60–78	67	Fig. 7.6a
Noble et al. (1994)	87	NR	66	Fig. 7.6a
Noble et al. (1997)	88	NR	70	Fig. 7.11b
Noble et al. (1998)	9	NR	69	Figs. 7.8, 7.11a, b, 7.12a
Picou et al. (2014)	18	48–83	69	Fig. 7.10b
Simon (2005)	≤5	NR	NR	Fig. 7.9a
Smith-Olinde et al. (1998)	6	32–64	NR	Fig. 7.6a
Vaillancourt et al. (2011)	≤57	25–61	48	Figs. 7.6a, 7.8, 7.12a, b
van den Bogaert et al. (2006)	10	44–79	NR	Figs. 7.6a, 7.8, 7.10b
van den Bogaert et al. (2011)	12	NR	NR	Figs. 7.6a, 7.8, 7.9b, 7.11a, b, 7.12a
van Esch et al. (2013)	72	22–91	63	Fig. 7.6b

If the number of subjects includes "≤" then the number of listeners varied according to condition or analysis. HI, hearing-impaired; NR, not reported

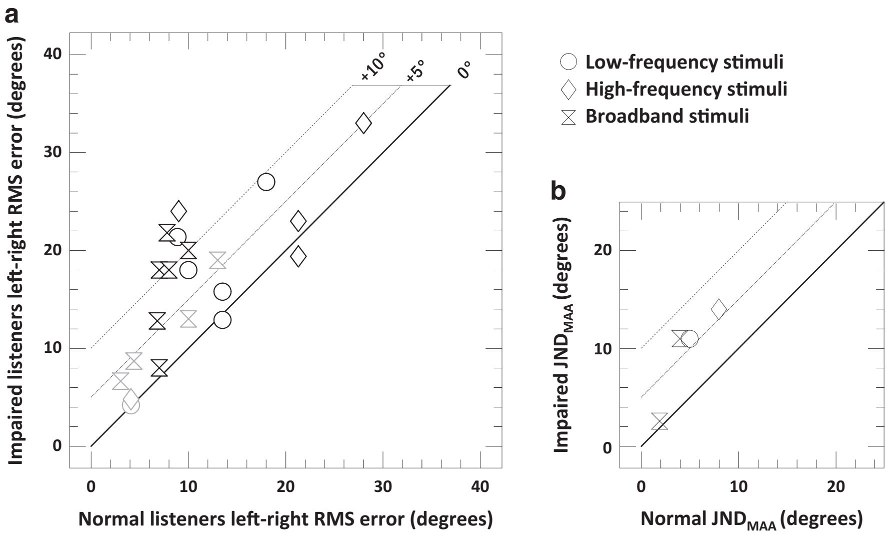

Fig. 7.6 (**a**) Summary of experimental data on the overall effect of hearing impairment on acuity in azimuth. There is one data point for each result reported in the various experiments (see Table 7.2 for the list). The symbols distinguish the frequency content of the stimuli: low frequency (*circles*), high frequency (*diamonds*), or broadband (*hourglasses*). The *gray symbols* indicate values converted from fraction correct or absolute error to RMS error. The *solid line* marks equality of performance for normal and impaired hearing; the *dashed lines* mark poorer performance for the hearing impaired by +5° and +10°. (**b**) The same analysis applied to minimum audible angle data

either equal to that for the normal group or worse by 5° or 10°. The symbols indicate whether the stimulus was low frequency (<1,500 Hz), high frequency (>1,500 Hz), or broadband. The broadband stimuli were mostly speech or noise but included recordings of environmental sounds such as a telephone ring or a cockatoo's call. All sounds were presented in quiet. Results reported as something other than *D* and converted to *D* here are shown in gray.

In general, the mean localization performance of the hearing-impaired/older listeners was slightly worse than for the normal-hearing listeners. The mean values of D_N, calculated across all the results, for the normal-hearing and hearing-impaired/older listeners were, respectively, 9° and 14°. The hearing impairment/age thus resulted in an overall deficit of 5°. The within-experiment difference between the groups was statistically significant in most, although not all, of the studies.

Two other experiments compared $\mathrm{JND}_{\mathrm{MAA}}$ values for mostly young normal-hearing and hearing-impaired/older listeners (Häusler et al. 1983; van Esch et al. 2013). Their results are shown in Fig. 7.6b. The mean values of $\mathrm{JND}_{\mathrm{MAA}}$ were, respectively, 5° and 11°. These were smaller than the corresponding D_{N} values, although at 6° the size of the group difference was essentially the same.

The distribution of differences in *D* between groups is shown in Fig. 7.7a. The majority of differences were positive, showing worse performance for hearing-impaired/

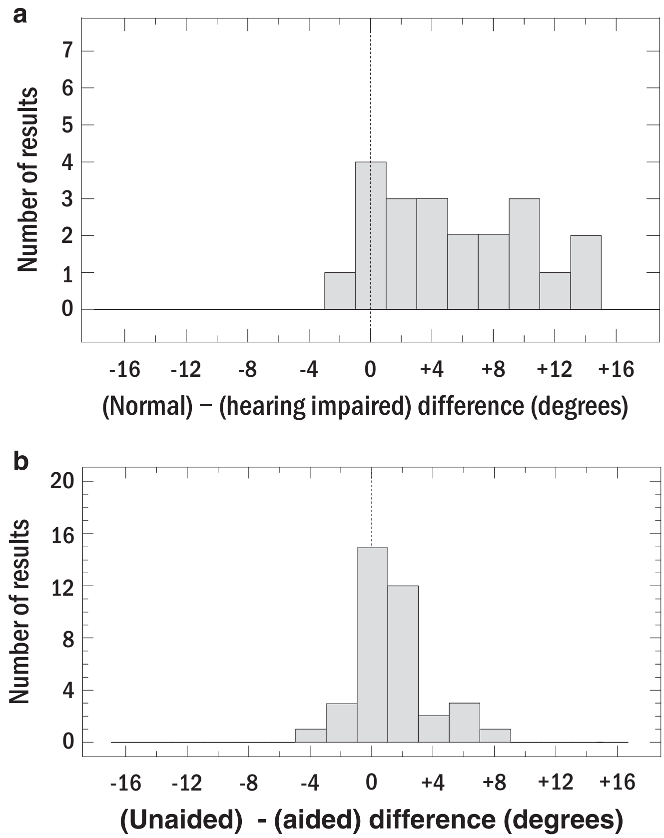

Fig. 7.7 (**a**) The distribution of within-experiment differences in acuity between normal and impaired hearing; positive indicates hearing impaired is worse. The *vertical line* marks no difference. (**b**) The same analysis applied to the differences between aided and unaided acuity; positive indicates aided is worse

older listeners. The range was −2° to 15°. Six of the differences were 10° or more: the studies of Keidser et al. (2009) using high-frequency noise and two kinds of broadband noise, Neher et al. (2011) using high-frequency noises, and Lorenzi et al. (1999b) using low-pass filtered clicks and broadband clicks.

Presumably the variety of methods, loudspeaker locations, stimuli, and listeners is largely responsible for the range of results shown in Figs. 7.6 and 7.7. It is arguable that this diversity means that the overall results can be taken as fairly representative of general performance for typical hearing-impaired listeners listening to a range of stimuli. However, there are at least three reasons to question how generalizable it is. First, all the results are for signals presented in quiet; localization is worse for sounds presented in noise if the signal-to-noise ratio is low enough (see Lorenzi et al. 1999a, b for psychometric functions down to a signal-to-noise ratio of −9 dB).

Second, the mean hearing loss of the hearing-impaired/older group was about 40–45 dB in almost every study, excepting that of Smith-Olinde et al. (1998) where it was 28 dB. Third, none of the experiments used age-matched controls. For the subset of experiments that reported mean or median group ages, the average age of the normal-hearing groups was about 30 years, while that of the hearing-impaired groups was 65 years (the other experiments reported ranges that mostly include these means). One exception was Abel and Hay (1996), who found a 4 % decrement in percent correct for older hearing-impaired listeners (age 4–73) compared to older normal-hearing listeners (41–58). However, all scores in their study were within 5 % of ceiling.

Further data on the effect of age were presented by Whitmer et al. (2014). They reported a set of statistically controlled correlations between localization precision, age, and hearing loss ($n = 35$, age = 26–81 years, hearing losses = −1 to 67 dB). The sensation level of the stimuli was 20 dB for all participants. They found a significant correlation of precision with age after partialing out the effect of hearing loss ($r = 0.46$) but not of precision with hearing loss after partialing out age ($r = 0.21$). These results, along with the effects of age on ITD processing summarized in Sect. 7.2, suggest that age per se is an important contributor to the group differences shown in Fig. 7.6.

Based on results for the six experiments that directly compared performance across stimulus types using within-listener designs, acuity for broadband stimuli was about the same as for low-frequency stimuli, and both gave higher acuity than high-frequency stimuli. The values of D_N were 1° lower for broadband stimuli than for low-frequency stimuli and 5° lower for low-frequency stimuli than for high-frequency stimuli. The difference between high-frequency and other stimuli was statistically significant for four of the six studies.

These data can be compared with those of Yost et al. (2013) for 45 normal-hearing listeners, aged 21–49 years, obtained using a fixed loudspeaker array. Their mean D value for a broadband stimulus (125–6,000 Hz noise) was 6°, which is comparable to the overall average of 8° found here. The changes in D produced by low- or high-pass filtering the stimuli (at 500 or 1,500 Hz, respectively) were less than 1°. The comparison suggests that the deficit in hearing-impaired/older listeners relative to normal-hearing listeners is greater at high than at low frequencies.

Only a few studies have reported correlations between an overall measure of hearing loss and localization acuity for broadband stimuli. Byrne et al. (1992) reported correlations of 0.11, 0.21, 0.42, and 0.66 between the Noble/Byrne error score and hearing loss averaged across 0.5, 1, 2, and 4 kHz, for four separate listener groups. Kramer et al. (1996) reported a correlation of −0.65 between percent correct and hearing loss across the same frequencies. A reanalysis of the data of Akeroyd and Guy (2011) (see Figs. 7.3 and 7.4) gave a correlation of 0.44 between D and hearing loss, and a reanalysis of the data of Whitmer et al. (2014) for click trains in quiet gave a correlation of 0.22 between D and hearing loss. Given the variety in methods, conditions, and results, the overall value of the correlation between hearing loss and localization acuity cannot be estimated with certainty but is probably close to the average of 0.4.

Correlations of localization acuity with hearing losses at individual audiometric frequencies were reported by Noble et al. (1994). For pink-noise stimuli presented at the most comfortable level, the correlation of the Noble/Byrne error score and hearing loss decreased from 0.42 at 0.25 kHz to 0.25 at 8 kHz. Noble et al. (1998) reported that the correlation of percent correct with hearing loss at 0.25 and 0.5 kHz varied from −0.74 to −0.09 across various experimental conditions. The only report of a correlation between left–right RMS errors and hearing loss for a high-frequency stimulus (4–8 kHz) gave a value of just 0.11 (Neher et al. 2011).

Given that there is some correlation between hearing loss and localization acuity, one should be able to derive the regression equation relating the two. Its parameters are, however, very uncertain. For the data of Akeroyd and Guy (2011) for words in babble (see Fig. 7.4c), the best-fitting straight line has a slope of 0.34°/dB (note that the choice of a straight line is based on an assumption that itself may be wrong). In contrast, for the data of Whitmer et al. (2014) for click trains in quiet, the slope of the regression equation is only 0.03°/dB. These two experiments differed in the presence or absence of a background sound and the type of target sound, but there may be other reasons to account for the considerable difference in slopes.

In summary, when pooled across these somewhat heterogeneous experiments, the acuity for changes in azimuth is about 5° poorer for hearing-impaired/older listeners than for (young) normal-hearing listeners. The difference has usually been statistically significant within experiments. Localization acuity is slightly worse with high-frequency stimuli than with low-frequency or broadband stimuli. There is a small to moderate correlation between localization acuity and hearing loss, with a mean value of roughly 0.4. The slope of the regression line relating localization acuity to hearing loss is even more uncertain. There is much that still needs to be done experimentally to determine the correlation and regression parameters between hearing loss and directional acuity.

7.4.3 Aiding and Left–Right Acuity

Figure 7.8 shows the data from the 11 studies that have compared the directional acuity of unaided and bilaterally aided hearing-impaired (mostly older) listeners. All studies used within-listener designs, but there was a wide diversity in aids and stimuli. Best et al. (2010) used BTE aids and completely-in-the-canal (CIC) aids; Byrne et al. (1996, 1998) used open or closed earmolds; Drennan et al. (2005) used aids that did not preserve IPD (another condition, using special aids that did preserve IPD, is not included here); Keidser et al. (2009) and Kuk et al. (2013) used BTE aids; Noble et al. (1998) used experimental earmolds that were closed, open, or sleeved, the latter being thin tubes that were even more open than a standard open fit; van den Bogaert et al. (2011) used in-the-pinna (ITP) or in-the-canal (ITC) hearing aids; and the studies of Vaillancourt et al. (2011), van den Bogaert et al. (2006), and Brungart et al (2014) used the listeners' own aids, with a variety of styles. The stimuli included speech, noises with various bandwidths, center frequencies, and

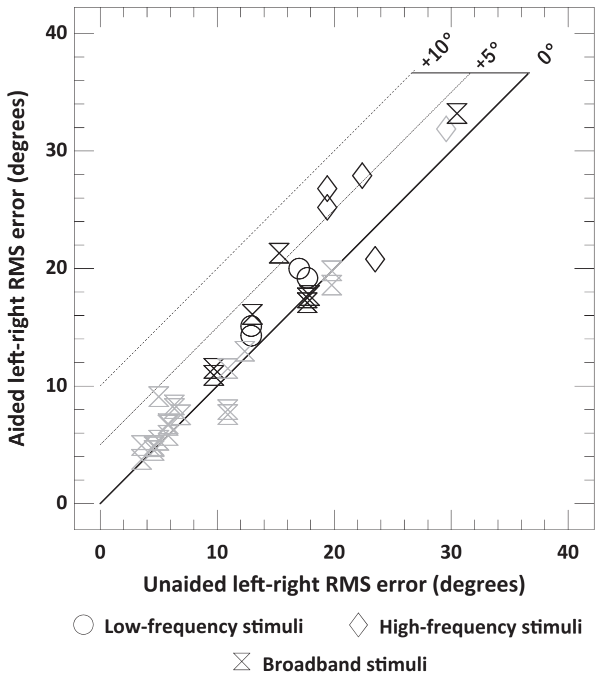

Fig. 7.8 Summary of experimental data on the effect of aiding on left–right acuity, plotted in the same format as Fig. 7.6

spectral slopes, and recordings of real-world sounds such as a telephone ring. Some sounds were presented in quiet and some were partially masked by background sound (see Brungart et al. 2014). The range of levels was 50–70 dB SPL. Taken together, this multiplicity likely allows the overall results to be interpreted as representative, although within the restrictions mentioned in Sect. 7.4.2. Data were excluded if they were obtained with unilateral fits (see Sect. 7.4.4 for these) or with hearing aids employing directional microphones (Sect. 7.4.7).

Across all the 36 results shown in Fig. 7.8, the mean values of D_N were 12° for unaided listening and 13° for aided listening, a tiny difference. Within experiment, the unaided–aided differences were found to be statistically significant by Keidser et al. (2009) and van den Bogaert et al. (2011) but not by Best et al. (2010), Drennan et al. (2005), or Brungart et al. (2014). For aided and unaided listening, the values of D_N were 15° and 17°, respectively, for low-frequency stimuli (four results), 23° and 25° for high-frequency stimuli (five results), and 10° and 11° for broadband stimuli (27 results).

The distribution of unaided–aided differences shows a trend toward worse performance when aided, although the difference was less than ±2° for approximately two-thirds of the results (see Fig. 7.7b). Only four results showed a *benefit* of aiding of at least 1°, the largest being 3° reported by Noble et al. (1998) and Keidser et al. (2009), whereas more than 20 results showed a deficit of aiding of at least 1° and nine (=25 %) of the results showed a deficit of 3° or more. The largest deficit was 7° (van den Bogaert et al. 2011).

Overall, the evidence suggests that hearing aids do not give any benefit for left–right directional acuity. Instead, they make performance worse by about 1° on average, though this would probably not be noticeable in everyday listening.

7.4.4 Unilateral Versus Bilateral Aiding and Left–Right Acuity

Figure 7.9a shows the results of the four studies that compared sound localization for unilateral and bilateral aiding. Only one of these (Keidser et al. 2007) reported *D* values; the others reported something else and scores were converted to *D*. All used between-listener designs with some form of broadband stimulus.

On average, performance was worse for unilateral aiding than for bilateral aiding; the values of D_N were 16° and 11°, respectively. Two of the experiments gave particularly large deficits for unilateral aiding: Keidser et al. (2007) found a statistically significant difference of 17° for a condition with 0-dB gain and with all the special features of the hearing aids turned off, and Byrne et al. (1992) found a statistically significant difference of 18° for a group of listeners with hearing losses greater than 50 dB fitted with BTE aids (these two results are indicated by arrows in Fig. 7.9a). Such effects would probably be easily noticed in everyday life. In con-

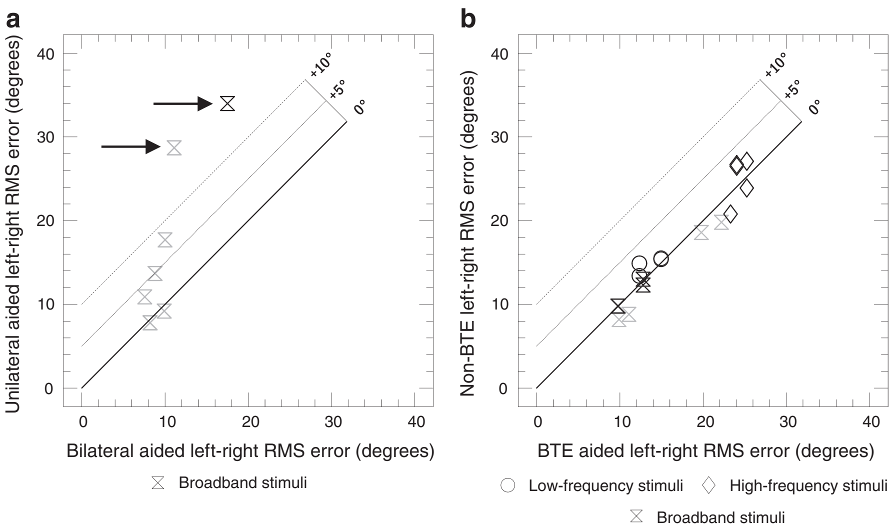

Fig. 7.9 (**a**) Summary of experimental data on the effect of unilateral versus bilateral fitting on acuity in azimuth. The *arrowed points* are highlighted in the text as they show particularly poor unilateral acuity. (**b**) Summary of experimental data on the effect of form factor on acuity in azimuth

trast, however, Byrne et al. (1992) did *not* find significant effects of unilateral versus bilateral aiding for listeners with greater than 50 dB loss fitted with ITE aids and for listeners with hearing losses less than 50 dB fitted with either BTE or ITE aids.

An experiment of Noble and Byrne (1991) also compared unilateral and bilateral aiding but was not included in Fig. 7.9a because of a confound in their measure of performance. Across three types of hearing aids (BTE, ITE, ITC) and three groups of hearing-impaired listeners (those already using BTE, ITE, and ITC aids), they found a mean localization error score of 1.8 per trial for bilateral aiding and 3.7 per trial for unilateral aiding. A higher error score is worse, as the nominal unit for localization error score was the number of loudspeakers by which the response differed from the target. Unfortunately, the error score cannot be converted directly to an acuity in degrees as its calculation used a mix of differences and products of mislocations as well as a combination across horizontal and vertical acuity. The data of Köbler and Rosenhall (2002) on source identification using eight loudspeakers in a full circle were also excluded from Fig. 7.9a because left versus right and front versus back errors were not distinguished. Again, there was a deficit in unilateral aiding: scores were 49 % with bilateral aids, 43 % with unilateral aids fitted to the better ears, and 41 % with unilateral aids fitted to the worse ears.

All of these data indicate that unilateral aiding can lead to a deficit in directional acuity, although given the mix of significant and insignificant effects reported by Byrne et al. (1992), it cannot be assumed that *everyone* will gain benefit from two aids. Many people are fitted with one aid, and two aids are more expensive (and require more clinical time to fit) than one. It is therefore somewhat disappointing that it remains unclear which individuals will benefit from bilateral versus unilateral aiding.

7.4.5 Form Factor and Left–Right Acuity

Four experiments have compared sound localization for bilaterally fitted BTE hearing aids to that for CIC aids (Best et al. 2010; Jensen et al. 2013), ITE aids (Byrne et al. 1992; van den Bogaert et al. 2011), ITP aids (van den Bogaert et al. 2011), and ITC aids (van den Bogaert et al. 2011). Figure 7.9b shows the 17 results, most of them from van den Bogaert et al. (2011), who tested four form factors using one group of listeners with a variety of stimuli. van den Bogaert et al. (2011) also used open molds and receiver-in-the-ear types of BTE. The data from these were averaged here as the degree of openness of the receiver-in-the-ear aids was unspecified.

On average, there was no difference between BTE aids and all the others taken together: the average values of D_N for both were 17°. Only van den Bogaert et al. (2011) found a benefit of any custom form over BTE, but none of the relevant differences in that experiment were statistically significant. Best et al. (2010), Byrne et al. (1992), and Jensen et al. (2013) all reported slight deficits for the custom form compared to BTE of about 1–2°, but again none of the deficits was statistically significant. Thus, overall, there is no evidence that form factor has an effect on localization acuity.

7.4.6 Acclimatization and Left–Right Acuity

Acclimatization, accommodation, or adjustment effects are well-known in hearing aid research (e.g., Arlinger et al. 1996; Munro 2008), although they are by no means ubiquitous (e.g., Humes and Wilson 2003; Dawes et al. 2014). Four experiments have considered the effects of acclimatization on directional acuity using mostly older hearing-impaired-adults. Figure 7.10a shows a summary of the data, plotting D for each condition as a function of time. Best et al. (2010) tested participants 0 and 4–6 weeks after fitting with CIC and BTE hearing aids, and Drennan et al. (2005) tested 0, 3, 13, and 16 weeks after fitting with IPD-preserving and nonpreserving hearing aids. Two studies by Keidser et al. tested hearing aids with various kinds of directional microphones and/or noise reduction: Keidser et al. (2006) tested at 2 weeks and 2 months (here taken as 8 weeks), and Keidser et al. (2009) tested at 0 and 3 weeks.

If a reliable general effect of acclimatization existed, then all the lines in Fig. 7.10a would slope downward to the right. Except for the first two points (0–3 weeks) of Drennan et al. (2005), this did not occur: the across-result values of D_N measured preacclimatization (measured 0 or 2 weeks after fitting) and postacclimatization (averaged across 3, 8, 13, or 16 weeks) were 20° and 19°, respectively. The effect is therefore minimal.

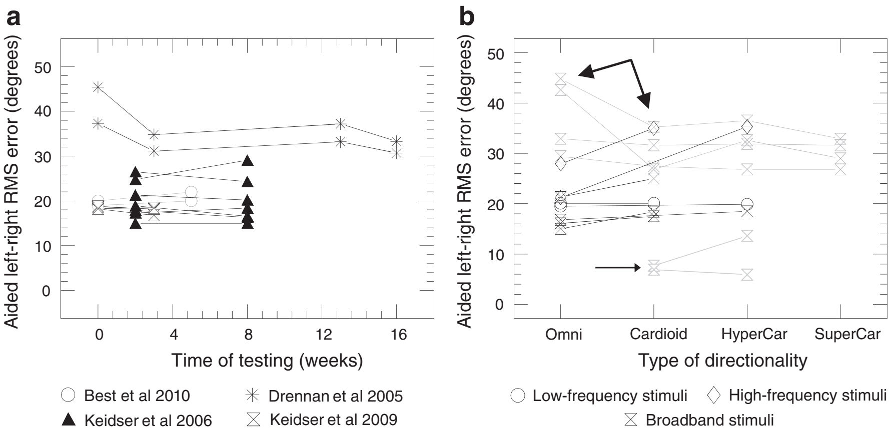

Fig. 7.10 (**a**) Summary of experimental data on the effect of acclimatization on acuity in azimuth. In contrast to the earlier plots, the symbols indicate the four experiments with one symbol per condition reported. The data of Best et al. (2010) obtained at 4–6 weeks are plotted at 5 weeks (their data were also converted from absolute error to RMS error). Keidser et al. (2009) used several stimuli, but only the data obtained using speech are presented here. (**b**) Summary of experimental data on the effect of using directional microphones on localization in azimuth. The directionality axis assumes the order: omnidirectional<cardioid<hypercardioid<supercardioid. The symbols distinguish the frequency content of the stimuli. The *arrowed points* are highlighted as they show a benefit of cardioid microphones or particularly good overall performance

7.4.7 *Directional Microphones and Left–Right Acuity*

Figure 7.10b shows the data from five experiments on the effect of directional microphones, assuming an ordering of the amount of directionality based on the ratio of responses for the front and rear hemifields: omnidirectional<cardioid< hypercardioid<supercardioid. The “adaptive directional” microphones used by van den Bogaert et al. (2006) are here classified as cardioid, and the “moderate” and “strong” directional processing modes of Picou et al. (2014) are classified as cardioid and hypercardioid, respectively.

If there were a reliable effect of increasing directionality, then all the lines in Fig. 7.10b would slope upward to the right if it was deleterious or downward to the right if advantageous. A large deleterious effect, 11°, occurred for high-frequency stimuli, plotted as diamonds. These results are from Keidser et al. (2009) and van den Bogaert et al. (2006). Note that not all experiments gave statistically significant results. No effect occurred for low-frequency stimuli (plotted as circles; overall difference <1°). Ignoring for the moment the two sets of data plotted in gray, the data for broadband stimuli also gave minimal effects, with an overall difference of about 1°.

The data of Chung et al. (2008), which are at the top of Fig. 7.10b in gray and marked by a pair of thick arrows, deserve further comment. Two of their conditions involved presentation from behind and gave a substantial *benefit*, on average 14°, for a cardioid microphone relative to an omnidirectional microphone. The effect was presumably due to the direction-dependent level differences introduced by the directional microphones (as applies to many of the results for directional microphones), and the differences were statistically significant. Their other two conditions involved presentation from the front, and for these the change in acuity from omnidirectional to cardioid was less than 1° and was statistically insignificant. The values of D for the three types of directionality across all four conditions changed by no more than 2° from cardioid through hypercardioid to supercardioid. It is not clear why the D values were relatively high compared to those for the other experiments (an average of about 40°); they used virtual-acoustics methods to present their stimuli, and perhaps it was a result of the use of manikin HRTFs rather than individualized HRTFs in the virtualization.

The other set of gray points, at the bottom of Fig. 7.10b and marked by the thin arrow, is from Picou et al. (2014). The effect of directionality was statistically significant in their results, but it interacted with loudspeaker location, such that the detrimental effects of directionality were present for targets presented from azimuths of ±60° but not ±45°. The overall values of D were far lower than for the other studies. This may be due to the transform used here to convert their published data from fraction correct to RMS error. The transform was based on data collected using a 13-loudspeaker arc, from −90° to +90° at 15° intervals, whereas Picou et al. used just four loudspeakers, at azimuths of ±60° and ±45°. Picou et al. also used a moderately reverberant room and stimuli presented at an SNR of 7 dB in babble, although if anything these factors would be expected to make localization performance worse rather than better.

Overall, the results indicate that the effects of directional microphones on azimuthal acuity are minimal except for high-frequency stimuli or stimuli presented from behind.

7.4.8 Front–Back Acuity

Figure 7.11a shows the effect of aiding on *D* for sounds presented from the side from five within-listener studies. The majority of conditions used broadband stimuli (hourglass symbols), but Keidser et al. (2009) also used a low-frequency noise (plotted as a circle) and a high-frequency noise (diamond). The data are for BTE aids (Keidser et al. 2009), ITC, ITP, and two forms of BTE aids (van den Bogaert et al. 2011), and BTE aids with open, closed, or sleeve molds (Byrne et al. 1996, 1998; Noble et al. 1998). Keidser et al. used a 360° circle of loudspeakers, with the present data derived by decomposing the results into the front-to-back dimension; the others used a sideways-placed 180° arc of loudspeakers.

For the broadband stimuli, the results show a small overall detrimental effect of hearing aids: the mean values of D_N for unaided and aided listening in all the conditions with broadband stimuli were 25° and 27°, respectively. There was a smaller

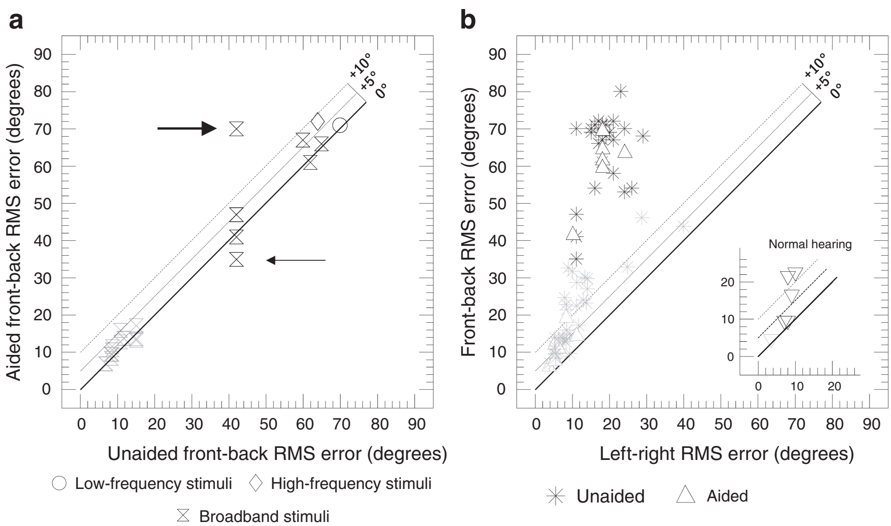

Fig. 7.11 (**a**) Summary of experimental data on the effect of aiding on front–back acuity using sounds presented from the side. The *arrows* mark two BTE conditions that gave a slight benefit or a substantial deficit in acuity. Note that the range of the axes is twice that in the earlier figures. (**b**) Summary of experimental data comparing left–right acuity with front–back acuity. The symbols indicate if the results are for aided or unaided conditions. The *main panel* plots data for hearing-impaired listeners, and the *inset panel* plots data for normal-hearing listeners to the same scale

difference, 1°, for the sole result using low-frequency stimuli, while the difference was substantially larger, 7–8°, for the two conditions with high-frequency stimuli.

Most of the other results included in Fig. 7.11a showed small effects of hearing aids, from ±2° to +5°. But there is an individual data point, marked by a thick arrow in Fig. 7.11a, that showed exceptionally poor aided performance, nearly 30° poorer. This condition also gave a very high increase in front–back confusions (see the arrowed point in Fig. 7.12a below). It is from van den Bogaert et al. (2011) and was obtained with a BTE device with its receiver in the ear, an omnidirectional microphone (though not perfectly so, as it was "front focused") and communication between the two hearing aids used to "improve the preservation of the interaural cues". Another BTE device using an open-fitting and a directional microphone gave a small *benefit* (marked by a thin arrow). Given that the deficit from the former device was the largest observed in any study surveyed in this chapter and would undoubtedly be noticeable in everyday life, it would seem important to follow up this result.

Figure 7.11b shows how front–back acuity relates to left–right acuity. It has been known for at least 75 years (e.g., Stevens and Newman 1936; Mills 1958) that localization acuity for sounds presented from the side is materially worse than for sounds presented from the front, even for normal-hearing listeners. For instance, Mills reported JND_{MAA}s of about 1° for a reference azimuth of 0° and 8° for a reference at

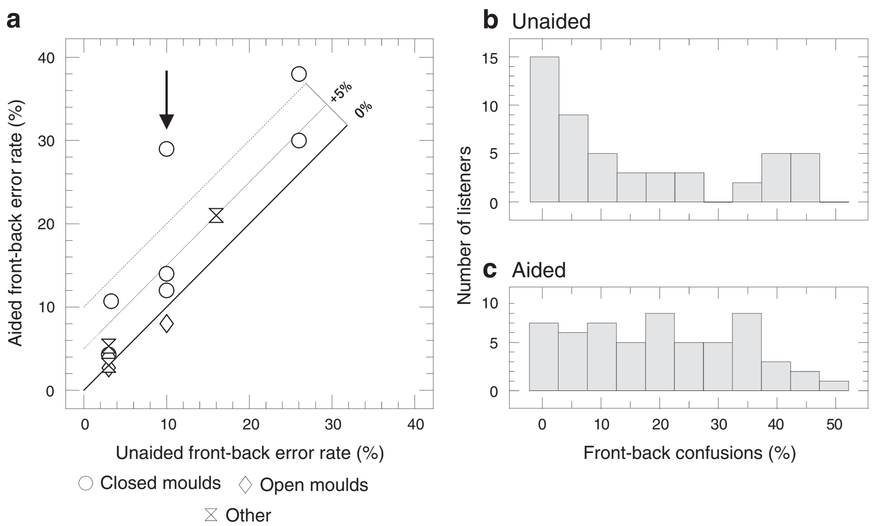

Fig. 7.12 (**a**) Summary of experimental data on the effect of aiding on front-back confusion rate. The symbols distinguish the earmolds used. The *arrow* points to the same condition as the one in Fig. 7.11a that gave very high aided confusions. (**b**) Distributions of front–back confusion rates for hearing-impaired listeners, tested unaided, from Vaillancourt et al. (2011, Fig. 7). (**c**) The corresponding distribution when tested aided. There are data for 50 individuals in panel **b** but 57 in panel **c**, as seven data points for unaided listening were hidden in the figure of Vaillancourt et al

75° (three normal-hearing listeners, 500-Hz pure-tone signal), and Häusler et al. (1983) found JND_{MAA} values for sounds presented from the front or side of 2° and 8°, respectively (36 normal-hearing listeners, 250–10,000-Hz white noise). The small inset panel in Fig. 7.11b shows *D* values for normal-hearing listeners from Byrne et al. (1996) and Keidser et al. (2009). The poorer front–back acuity is evident by most of the points being above the 1:1 line. The main panel shows the front–back versus left–right *D* values for hearing-impaired listeners using within-listener, within-stimulus designs. The symbols indicate if the data were collected in unaided (asterisks) or aided (triangles) conditions. The data cluster far above the 1:1 line. The best-fitting linear regression is $y = 2.5x + 4.2$ ($r^2 = 0.56$). That is, front–back acuity is, on average, about 2.5 times poorer than left–right acuity. Furthermore, a comparison to the inset panel (which is drawn to the same scale as the main panel) clearly indicates that hearing-impaired listeners, be they unaided or aided, have far poorer front–back acuity than normal-hearing listeners. This is further demonstrated by Häusler and colleagues' (1983) JND_{MAA} data for sounds presented from the side (cf. Fig. 7.4b). The median result was 8° for 6 normal-hearing listeners and 12° for 14 bilaterally hearing-impaired, older listeners with good speech discrimination ability. Even the best result for a second group of impaired listeners, with poor speech discrimination, was 15°, and more than two-thirds of that group scored 30°, this being the largest value that the method of Häusler et al. could measure.

7.4.9 Front–Back Confusion

Figure 7.12a shows a summary of the front–back confusion data from four experiments that compared aided versus unaided listening using within-listener designs. The stimuli were all broadband sounds, either noises or speech. Noble et al. (1998), Vaillancourt et al. (2011), and van den Bogaert et al. (2011) used a loudspeaker array to the side of the listener, whereas Best et al. (2010) used a single loudspeaker on a movable boom (these last data are also averaged across pre-/postacclimatization). The different symbols in Fig. 7.12a represent the types of hearing aid mold used: closed, or at least occluded, molds, including CIC, ITC and ITP aids (circles); open molds (diamonds); and various other types (hourglasses).

The results show that aids with open molds did not change the rate of front–back confusions compared to unaided listening: averaged across studies the unaided–aided difference was −1 %. Aids with closed molds, however, increased the rate compared to unaided listening (+6 % difference). The arrow marks the condition from van den Bogaert et al. (2011) that was highlighted earlier, as it gave a high front–back RMS error. It also gave a high front–back confusion rate.

There is substantial individual variation across listeners. Figure 7.12b, c illustrates the distribution of unaided and aided confusion rates from Vaillancourt et al. (2011) for up to 57 listeners, divided in 5 % wide bins. The scores go from perfect (0 %) to chance (50 %), and there are more perfect scores for unaided than for aided listening. Large individual variations for both unaided and aided listening were also reported by Best et al. (2010) (e.g., unaided range 3–40 %, 11 listeners) and van den Bogaert et al. (2011) (unaided range 0–32 %, 13 listeners).

There is also a large variation in unaided confusion rates across studies. The differences may simply be due to averaging the results from relatively small numbers of particularly diverse individual results, but it is worth considering other aspects. Stimulus duration would be expected to have some effect; if the stimulus is long enough, listeners could use head movements to help resolve front–back confusion. This might help account for the low confusion rate of 3 % observed by Noble et al. (1998), who used a 1-s duration, whereas van den Bogaert et al. (2011) used a 200-ms duration (10 % confusion rate) and Vaillancourt et al. (2011) used a 250-ms duration (16 % confusion rate). However, duration fails to explain the high confusion rate (26 %) reported by Best et al. (2010), who used monosyllabic words. The remainder of the difference may be due to Best et al. (2010) using words, whereas the others used noise, but the reasons remain unclear.

There are some reports that the confusion rate is correlated with hearing loss. Vaillancourt et al. (2011) found that the rate of confusions was moderately correlated ($r = 0.5$) with the average hearing loss across 3,000, 4,000, and 6,000 Hz; van den Bogaert et al. (2011) reported a significant correlation between performance and hearing level but did not give its value; Noble et al. (1998) found a substantial and significant correlation ($r = 0.84$) between front–back confusion rate and low-frequency hearing loss (250/500/1,000 Hz) in one of their conditions. On the other hand, Best et al. (2010) reported a correlation of just 0.14 between the rate of confusions and high-frequency hearing loss (4,000/6,000/8,000 Hz). Overall, therefore, the quantitative relationship between hearing level and the rate of front–back confusions is uncertain, and much of the variation across individuals remains unaccounted.

Two recent studies compared front–back confusion rates for normal-hearing listeners to those for unaided, mostly older, hearing-impaired listeners. Best et al. (2010) found mean rates of 5 % for normal and 26 % for hearing-impaired listeners; van den Bogaert et al. (2011) found mean rates of 0.1 % for normal and 10 % for hearing-impaired listeners.

In summary, normal-hearing listeners make a few front–back confusions, but older hearing-impaired listeners are particularly prone to them. Open-mold hearing aids do not change the rate of front–back confusions compared to unaided listening, but closed-mold hearing aids further increase it by, on average, 6 %.

7.5 Summary

Data on the effects of hearing impairment indicate that, on average, the acuity for left–right changes in azimuth is about 5° worse for unaided hearing-impaired mostly older listeners than for normal-hearing, mostly younger listeners. While not a particularly large effect, within study it was usually statistically significant, and it is probably large enough in magnitude to be meaningful. It corresponds to an azimuth difference of about half the subtended width of one's hand held at arm's length. There may be a small-to-moderate correlation, of uncertain size but perhaps about 0.4, between acuity and hearing loss. There are large variations in individual performance: some hearing-impaired individuals perform much more poorly than normal-hearing listeners, while others show as keen acuity. Many studies have compared younger normal-hearing listeners to substantially older hearing-impaired listeners, and there is evidence that age and hearing loss act separately.

The data for the effects of hearing aids show that, with the important exception of unilateral fitting, the effects are mostly minimal. There is no hearing aid feature or design that has unambiguously been shown to give a benefit over some alternative. Indeed, there are many statistically insignificant results mixed in with statistically significant ones. Although no formal statistical meta-analyses were done here, the data suggest that there is no reliable experimental evidence for a general, substantial benefit of hearing aids on directional hearing.

Specifically, the overall effect of bilateral hearing aids relative to unaided listening is a deficit of 1° in RMS error; the overall effect of directional microphones relative to omnidirectional microphones is a deficit of 3°; the overall difference of custom form factors relative to BTE is 0°; and the overall improvement from acclimatization is 1°. The results for low-frequency stimuli are much the same as those for broadband stimuli. For high-frequency stimuli, the aided-to-unaided deficit is 3°, while the BTE-to-custom difference is 0°. The measurement error in these values is likely about 1–2°, so there can be little confidence that any of these deficits are really different from zero. One exception is that directional microphones led to larger errors than omnidirectional microphones for high-frequency stimuli by 10°.

The one clear effect of aiding was a substantial deficit from unilateral relative to bilateral aiding. The acuity for unilateral fitting was, on average, 5° worse than for bilateral fitting, and some studies showed deficits close to 20°. If these results from the laboratory hold for everyday listening, the effects would certainly be noticeable. Nevertheless, even for this comparison, not every experiment showed a significant difference, and there is a need for more work in this area.

With regard to front–back discrimination, hearing-impaired listeners make far more front–back confusions and have poorer acuity than normal-hearing listeners. Some forms of hearing aid, such as closed molds, introduce yet further errors.

In conclusion, hearing aids do not offer any benefit for directional acuity in azimuth and they certainly they do not return it to normal. Nevertheless, only for unilateral fittings do hearing aids produce a large deficit. Two quotations serve to summarize all these results. Häusler et al. (1983) commented that "our results indicate that sound localization performance is not improved by wearing a hearing aid" (p. 400). Over 30 years later, there is little new experimental evidence to oppose this. Second, van den Bogaert et al.'s (2006) title, "Horizontal localization with bilateral hearing aids: Without is better than with" is still applicable, mostly. It could now be updated to "Horizontal localization with bilateral hearing aids: With is no worse than without, except sometimes at high frequencies, but unilateral fittings can be substantially worse."

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Chapter 8
Music Perception and Hearing Aids

Justin A. Zakis

Abstract The development of hearing aids has been focused on improving speech understanding in quiet and noise, while appropriate processing for music has been a secondary consideration. This may explain some of the user dissatisfaction with hearing aids when listening to music. Music has a greater dynamic range, bandwidth, and range of spectrotemporal properties than speech, which places greater demands on hearing aid circuits and processing algorithms. Although music is an important sound for hearing aid users, little research has investigated how well hearing aids handle music, and even less has investigated how hearing aids should handle music. This chapter provides a review of the limitations, efficacy, and requirements of hearing aids for processing music. It begins with an overview of the spectrotemporal properties of music. This is followed by a discussion of the ability of hearing aid circuits to handle the dynamic range and bandwidth of music without distortion. The perception of rhythm, pitch, melody, and timbre by hearing aid users is briefly reviewed. The literature on appropriate processing for music by a wide range of algorithms commonly found in hearing aids is discussed. The chapter concludes with a summary and future research directions.

Keywords Analog-to-digital converter • Bandwidth • Digital signal processor • Digital-to-analog converter • Dynamic range • Fitting • Microphone • Preamplifier • Processing algorithms • Receiver

8.1 Introduction

Historically, hearing aids have been designed and fitted with the primary goal of restoring speech intelligibility. Although understandable, this approach has often been at the expense of music sound quality. Listening to music is important to about three-quarters of hearing aid users, with around half listening to music daily (Leek et al. 2008; Kochkin 2010). About one-quarter of aid users believe their hearing loss interferes with their

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enjoyment of music (Leek et al. 2008) and about two-thirds believe their enjoyment has decreased since the onset of hearing problems (Uys et al. 2012). People with musical training—even mainly as children in primary school—are more likely to suffer reduced music enjoyment after developing a hearing loss (Leek et al. 2008). Fewer than half of hearing aid users find that their aids make music more enjoyable, while around 5–10% find that their aids actually make music *less* enjoyable (Leek et al. 2008) or are unsatisfactory when listening to music (Kochkin 2010). The latter is consistent with the percentages of people who find that hearing aids make listening to recorded music a bit or much worse, and these percentages increase for live music (Madsen and Moore 2014).

Common problem areas with hearing aids include understanding sung words, an overall volume that is too loud or soft, excessive short-term changes in loudness, and to a lesser extent melody recognition (Leek et al. 2008). Hearing aids can also make loud parts too loud, reduce clarity, worsen tone quality, make it harder to hear individual instruments, or make music seem distorted, too bright/shrill, or lacking in bass (Madsen and Moore 2014). The softness, nearness, and loudness of classical music can be further from ideal values at a characteristic peak level for live performance than at a lower listening level (Narendran and Humes 2003). There is clearly room for improvement in how hearing aids process music and probably more so for people with musical training and particularly practicing musicians, who may listen at higher input levels and have more demanding requirements (Zakis and Fulton 2009; Chasin and Hockley 2014) that can be better met with assistive listening devices (Revit 2009; Einhorn 2012).

There is also substantial *potential* for improved music processing because improvements can be made at multiple points in the signal path. This involves modifications to circuits and algorithms for the wider dynamic range, bandwidth, and range of spectrotemporal properties of music than of speech. The main problem with microphones and front-end circuits is handling the dynamic range of music without introducing significant amounts of clipping distortion or noise. Common prescriptions for wide dynamic range compression (WDRC) algorithms are tailored for speech inputs (Scollie et al. 2005; Moore et al. 2010; Keidser et al. 2011), and many hearing aid manufacturers modify the prescribed (e.g., gain-frequency response, compression ratios, and/or output limits) and/or nonprescribed (e.g., time constants) settings in different ways to improve sound quality in the music program. Although some differences could be expected as a result of different algorithm implementations and interactions, it is not clear from the limited amount of literature which general approach is best for music and whether this depends on factors such as differences in signal processing across aids, individual characteristics (e.g., hearing loss or cognitive factors), or the music (e.g., genre, instrument, or spectrotemporal properties). Adaptive dynamic range optimization (ADRO) may be preferred to WDRC for music with first-fit settings (Higgins et al. 2012), although adjustment of the ADRO fitting predictions can be beneficial for exacting clients such as musicians (Zakis and Fulton 2009). Other algorithms such as adaptive directionality, frequency lowering, and the reduction of feedback, reverberation, and various types of noise (see Launer, Zakis, and Moore, Chap. 4) may also need to be retuned or deactivated for music, although the relevant literature is thin to nonexistent. Output circuits and transducers (receivers) may not always be capable of delivering the bandwidth and high-frequency output levels required for music. All or even a subset of these potential improvements (if achievable) could add up to a substantial net improvement in the experience of listening to music through hearing aids.

This chapter reviews the main factors that limit the performance of hearing aids with music and some solutions. The review is mostly limited to conventional air conduction hearing aids.

8.2 Acoustic Characteristics of Music

The spectral and temporal characteristics of music can be similar or very different to those of speech. Singing is similar to speech because in both cases the vocal tract is used to generate sound. During voiced sounds, the vocal chords vibrate at a fundamental frequency (F_0) and generate harmonics at all integer multiples of the F_0. The harmonic spectra are shaped by the resonances of the damped vocal tract. During unvoiced sounds, the vocal chords do not vibrate and the vocal tract spectrally shapes a noiselike air flow (Johnson 2003). When the vocal tract is used for singing, its F_0s and temporal characteristics can cover a much wider range than during talking. Acoustic musical instruments have a wider range of vibrating sources (e.g., strings, reeds) and less damped resonators (e.g., wooden body, pipe) that result in an even wider range of F_0s, spectral shapes, and temporal properties. Electronic circuits with instruments such as the electric guitar can generate their own harmonics, while computer-generated music can potentially have any spectrotemporal characteristics.

Favored music genres vary with hearing aid users and across populations. A US survey showed that classical music was clearly the most popular genre, followed in roughly equal measures by oldies/rock, pop/easy listening, and jazz (Leek et al. 2008). In contrast, a South African survey ranked folk/country and classical as the most preferred genres and rock and jazz/blues as the least preferred, while pop, ballad singing, opera/operetta, choir, and dance music fell in between (Uys et al. 2012). These genres span a diverse range of instruments and playing styles that cover a wide range of spectral and temporal properties. Hearing aid users also listen to music from a variety of sources (e.g., radio, recorded media, live music, and television) and in diverse situations (e.g., in a car, house, or church) that modify the spectral and temporal properties of music in different ways (Leek et al. 2008). Sections 8.2.1 and 8.2.2 provide a brief overview of these properties compared with those of speech.

8.2.1 Spectral Characteristics

The long-term average speech spectrum (LTASS) is very well defined and fairly consistent across languages and accents from approximately 0.1–16 kHz (Byrne et al. 1994; Moore et al. 2008). Hearing aids typically amplify only the lower half of this range, which is adequate for good speech intelligibility in quiet (ANSI

S3.5, 1997). The LTASS results from the combination of the low-frequency weighted, harmonic spectra of voiced speech sounds and the high-frequency weighted, nonharmonic spectra of unvoiced speech sounds. These predictable characteristics facilitate the targeting of amplification settings to speech (as opposed to music) and the identification of speech as opposed to other sounds (often assumed to be noise) by various processing algorithms.

In comparison, the music spectrum is much less predictable and varies widely. The F_0 of the human voice can be two or three octaves higher for singing than for talking (Revit 2009). Musical instruments can generate F_0s as low as 31 Hz and as high as 4–5 kHz (Revit 2009). Different instruments can generate harmonics at all integer multiples, odd-numbered integer multiples, or noninteger multiples of the F_0 (Chasin and Russo 2004; Chasin 2006, 2012). Harmonic spectra can have low-, mid-, or high-frequency emphasis and can have significant harmonic energy up to 10 kHz (Revit 2009). Nonharmonic sounds include the interresonant lower frequency breathiness of a clarinet (Chasin 2006) and cymbals that can extend up to about 16 kHz (Revit 2009). Therefore, a typical music spectrum does not exist. This requires a more general-purpose approach to prescribing amplification settings and makes it challenging for processing algorithms to identify and hence process music appropriately.

8.2.2 Temporal Characteristics and Dynamic Range

Speech has a well-defined range of sound pressure levels. The long-term average level ranges from approximately 55–82 dB SPL at a distance of 1 m depending on vocal effort (Pearsons et al. 1977; ANSI S3.5, 1997), with instantaneous peak levels exceeding the average by about 20 dB (Cox et al. 1988; Byrne et al. 1994; Moore et al. 2008). Therefore, peak speech levels do not often exceed 100 dB SPL. Typical noise floor and clipping levels of hearing aid microphones and front-end circuits are well suited to the dynamic range of speech (see Killion, Van Halteren, Stenfelt, and Warren, Chap. 3).

Live music has a considerably greater range of sound levels than speech. Levels can be as low as 20–30 dB SPL (Clark and Luce 1965; Chasin 2006), which is similar to self-generated noise levels of typical hearing aid microphones (see Killion, Van Halteren, Stenfelt, and Warren, Chap. 3). Levels can peak at 125 dBA for classical music and even higher for rock music (Royster et al. 1991; Chasin 2006). As discussed in Sect. 8.3, such peaks may be clipped in linear hearing aid circuits or subjected to compression distortion by input and/or output limiting circuits. Music also has a wide range of other temporal properties, such as attack times that vary across instruments (Luce and Clark 1965), sustained notes, vibrato, and so forth, that can be very different from those of speech. As discussed in Sect. 8.5, the dynamic range and temporal properties of music place additional demands on amplification algorithms to keep music audible but not too loud and with minimal distortion. These properties may result in some other types of processing algorithm (e.g., noise reduction) treating music as an unwanted sound.

8.3 Hearing Aid Circuits

Figure 8.1 shows a simplified block diagram of the signal path of a generic digital hearing aid, which can be divided into six parts (from left to right): microphone, preamplifier (PA), analog-to-digital converter (ADC), digital signal processor (DSP), digital-to-analog converter (DAC), and receiver. Many hearing aids use additional preamplifier/ADC chains for a second microphone, telecoil (see Mecklenburger and Groth, Chap. 5), and direct audio input that are not shown here for simplicity. Hearing aids can also have multiple DSPs and coprocessors, which for simplicity have been lumped together as "the DSP" in Fig. 8.1 and for the remainder of this chapter. Briefly stated, the microphone converts sound pressure into an analog voltage waveform, which is amplified into the ADC's input range by a preamplifier. The ADC converts the analog voltage waveform to a discrete numerical value (sample) at regular time intervals and sends the samples to the DSP. The DSP runs algorithms that analyze and manipulate the samples to provide amplification and other features, such as noise reduction and feedback cancellation. The DAC converts the processed samples to voltages that a differential-drive receiver converts to sound pressure.

The DSP algorithms cannot be expected perform optimally unless the other circuits deliver audio signals with little (if any) added noise or clipping distortion. However, this is a challenge for single-supply circuits that must process audio signals over a low voltage range (about 0–1 V) with low power consumption (hundreds of μA) and low noise levels (the latter two are often traded off against each other). The requirements and limitations of each circuit for music are discussed in Sects. 8.3.1–8.3.6.

8.3.1 Microphones

The ideal microphone for music would handle peak levels without distortion, have nonintrusive self-generated noise levels during soft music passages, and have linear operation between these extremes. Electret microphones commonly used in hearing aids can handle input levels of up to 115 dB SPL with low distortion (Chasin 2012; see Killion, Van Halteren, Stenfelt, and Warren, Chap. 3), which is sufficient for all but the most extreme music peaks. However, this limit usually varies with frequency and differs across microphone models. In the author's experience, peak clipping is most likely to occur in a microphone's electronic circuits (hard clipping) and in some cases is due to the diaphragm entering a mechanical nonlinear region (soft clipping) that can occur for levels as low as

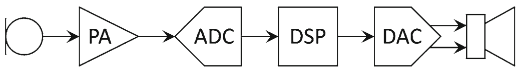

Fig. 8.1 Simplified block diagram of the audio signal path of a hearing aid, consisting of (from *left* to *right*) a microphone, preamplifier (PA), analog-to-digital converter (ADC), digital signal processor (DSP), digital-to-analog converter (DAC), and receiver

100 dB SPL. Peak clipping creates unpleasant harmonics that were not present in the source music. Even a small amount of clipping can reduce music sound quality for a person with impaired hearing (Tan and Moore 2008; Arehart et al. 2011). A higher clipping level can be achieved at low frequencies with a microphone that has a response that rolls off by 6 dB/octave below approximately 1 kHz (Schmidt 2012). However, low-frequency self-generated noise levels are increased from roughly 25 up to 45 dB SPL at 200 Hz (Schmidt 2012), which may require more aggressive low-level expansion algorithm settings that could potentially suppress soft music. Even microphones with a nominally flat frequency response typically have a wideband equivalent input noise level of about 30 dB SPL, which has the potential to exceed hearing threshold levels with some combinations of gain and hearing loss (Wise and Zakis 2008) and become audible during music pauses in quiet conditions (e.g., classical or jazz pieces in a quiet concert hall).

8.3.2 Preamplifier

A 16-bit ADC's samples represent a 96 dB range (6 dB per bit) that encapsulates the dynamic range of a typical hearing aid microphone and would appear to make a preamplifier unnecessary. However, in comparison with the 96 dB *numerical* dynamic range of the samples at the ADC's output, the ADC's *input* dynamic range may be reduced to only 80–85 dB to meet power consumption requirements (Ryan and Tewari 2009; Schmidt 2012). Power consumption and circuit noise are often traded off against each other, so the reduced input range is a result of increased noise levels that can exceed the microphone noise levels. Figure 8.2 (second row) shows an example where the microphone, preamplifier, and ADC have the same clipping level (upper solid line), but the ADC has a higher noise level (lower solid line). In this case, only the top part of the microphone's dynamic range will pass though the ADC's input dynamic range (dashed lines) and lower-level input signals will be lost. The third row of Fig. 8.2 shows that 15 dB of preamplifier gain are required to lift the microphone noise (and possibly also very soft music) above the ADC noise (as shown by the lower dotted line), which also reduces the equivalent input noise level. However, this gain will cause levels in the top part of the microphone's dynamic range (above the upper dotted line) to be clipped in the preamplifier, or if the preamplifier uses an input-limiting automatic gain control (AGC) to avoid clipping, the AGC's threshold in decibels SPL will be reduced. In both cases, either clipping or compressive limiting distortion occurs at a lower input level than with less preamplifier gain.

Thus, if a hearing aid uses a fixed preamplifier gain, a compromise is often made between a low gain for handling high-level music without distortion and a high gain for minimizing equivalent input noise and passing very soft sounds through the ADC's input bottleneck. One solution is to use a higher preamplifier gain for the normal hearing aid program to handle speech levels and a lower preamplifier gain for the music program to handle live music levels. Such a scheme allows music levels of up to 110 dB SPL to be handled with low distortion and

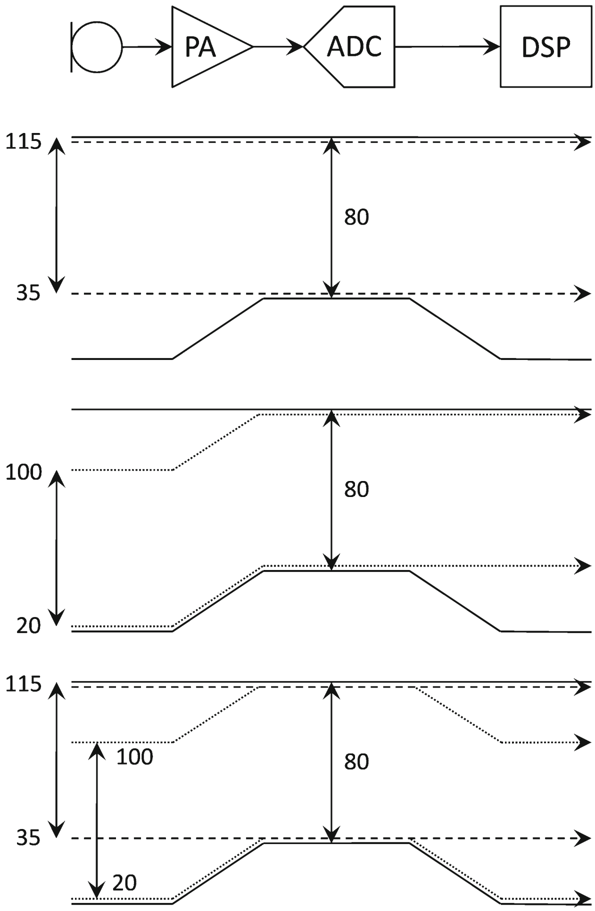

Fig. 8.2 Dynamic range of the microphone and analog-to-digital converter (ADC) circuits. The block diagram shows a high-quality microphone with a dynamic range that extends from 20 to 115 dB SPL, a preamplifier, an ADC (80 dB input dynamic range), and a digital signal processor (DSP). The *solid lines* in each row show the limits of these dynamic ranges (assuming the upper limit is the same for each circuit). The top row shows that with 0 dB preamplifier gain, input levels from 35 to 115 dB SPL pass through the dynamic range of the ADC (as shown by the *dashed lines*) while lower levels do not. The middle row shows that with 15 dB of preamplifier gain, input levels from 20 to 100 dB SPL pass through the dynamic range of the ADC undistorted (as shown by *dotted lines*), while higher levels are clipped in the preamplifier/ADC. The bottom row shows a system where the preamplifier gain is automatically adjusted for the input signal and then compensated in the digital domain, which allows input levels from 20 to 115 dB SPL to pass through the ADC and to the DSP

improves sound quality as judged by musicians (Hockley et al. 2012). This is consistent with a previous study where professional musicians rated peak input limiter thresholds of 105 and 115 dB SPL significantly better than 92 and 96 dB SPL (Chasin and Russo 2004). The side effect of increased equivalent input noise levels in the music program could be mitigated with more aggressive low-level expansion settings (Hockley et al. 2012), although these settings also have the potential to suppress very soft music.

Other solutions include dynamically adjusting the preamplifier gain to increase the input range to 96 dB (Ryan and Tewari 2009) and automatically switching between two ADCs connected to preamplifiers with different gains (for the same microphone) to increase the input range up to 110 dB (ON Semiconductor 2009). As shown in the bottom row of Fig. 8.2, the preamplifier gain is undone in the digital domain to preserve linearity of the microphone signal, so these two solutions combine high clipping levels and low equivalent input noise levels in one program.

8.3.3 Analog-to-Digital Converter

As discussed in Sect. 8.3.2, the 16-bit ADCs typically used in hearing aids have an input range of only 80–85 dB to meet power consumption requirements (Ryan and Tewari 2009; Schmidt 2012), and methods have been devised to shift this range into the desired range by adjusting preamplifier gain either statically or dynamically and compensating in the digital domain. The resultant sample values can extend over a range of more than 16 bits if the shifts are adjusted dynamically over a sufficiently wide range, effectively extracting more than 16 bits out of a 16-bit ADC. Alternatively, some hearing aids utilize a switched-capacitor voltage transformer to increase the ADC's dynamic range to 96 dB, so that input levels from 17 to 113 dB SPL are handled with low distortion, noise, and power consumption (Baekgaard et al. 2013).

It is important for the ADC's samples to have a sufficient number of bits to provide a dynamic range at least as great as the dynamic range of the microphone (or the music). Otherwise, very soft music may fall below the range of possible sample values (assuming the greatest sample value corresponds to the maximum ADC input). A reduced number of bits also increases quantization error (i.e., the rounding error when converting from an analog-to-digital value) and the resulting distortion and noise. Music needs to be represented by at least 7 bits to avoid reduced sound quality with linear amplification (Arehart et al. 2011) and this may not always be possible for the softest music sounds with 16-bit samples depending on the preamplifier gain and microphone sensitivity. An increased number of bits could be used to extend the range of sample values down to lower levels to better represent soft music. However, more bits also increases power consumption and hence reduces battery life.

Another key ADC parameter is its sampling rate. Compact disc (CD) or other digital media typically use a 44.1-kHz sampling rate. The Nyquist theorem states that this allows frequencies up to half the sampling rate (22.05 kHz) to be processed without frequency aliasing. This upper frequency limit is somewhat reduced by antialiasing filters, but it remains above the highest frequency of human hearing. Lower sampling rates are used in hearing aids because (1) higher sampling rates result in higher power consumption; (2) the high-frequency cutoff of conventional air conduction hearing aid receivers is much less than the 22-kHz cutoff of music media; (3) there is little research on appropriate high-frequency amplification for speech or music. In theory, a hearing aid with a 10-kHz input cutoff could use a 20-kHz sampling rate to avoid frequency aliasing. However, this does not ensure an accurate representation of the amplitude and phase of high-frequency spectral components. This may be one reason why some hearing aids use higher sampling rates, such as 33.1 kHz, to process a 10.5-kHz bandwidth (Baekgaard et al. 2013).

Hearing aids can use minimum-phase, low-pass, antialiasing filters to achieve a very steep roll-off with an ADC delay of less than 0.5 ms at 1 kHz (Ryan and Tewari 2009). It is unknown whether the phase-frequency response of such steep minimum-phase filters affects music perception or is preferred to a linear-phase filter that imposes a greater delay.

8.3.4 Digital Signal Processor

The DSP algorithms (Launer, Zakis, and Moore, Chap. 4) basically load samples and data into a small number of registers to perform mathematical operations and store samples, data, and results of these operations in memory banks. The registers and memory must have a sufficient number of bits to process and store the samples without accumulating significant amounts of rounding error that could lead to distortion (Ryan and Tewari 2009), and without creating a dynamic range bottleneck that can reduce music sound quality (Arehart et al. 2011). As a practical example, to run a 128-coefficient finite impulse response (FIR) filter to provide frequency-dependent amplification, a DSP needs to multiply 128 samples by 128 filter coefficients and accumulate the sum of these products. If the samples and coefficients are 16-bit values, then each product is 32 bits long, and the sum of the products will be 128 times larger and hence 39 bits long. Thus, the register that accumulates these products must be at least 39 bits long to avoid numerical clipping and/or truncation affecting the filtered audio signal. More bits translates to greater manufacturing cost and power consumption, so a number is chosen to meet foreseeable requirements.

Power consumption is why DSPs in hearing aids run at clock speeds in the megahertz to tens of megahertz range compared with personal computers that currently run in the gigahertz range. This severely limits the maximum number of computations that can be performed on each sample (proportional to the clock rate divided by the sampling rate) and hence the complexity, performance, and number of algorithms that can run in real time. It follows that higher sampling rates exacerbate this limitation because they reduce the number of computations that can be performed on each sample (or block of samples) before the next sample (or block of samples) arrives. Although music may require relatively high sampling rates, it may also be desirable to take a less-is-more approach and run fewer algorithms, although more complex versions of these algorithms (e.g., distortion-free acoustic feedback canceler) that use more processing time may also be required. Algorithmic issues are discussed further in Sect. 8.5.

8.3.5 Digital-to-Analog Converter

Class D amplifier circuits are typically used in digital hearing aids because of their low power consumption compared with DACs used in high-end CD players. The output samples are converted to a high-frequency (MHz range), pulse-width-modulated square wave that differentially drives the receiver. This frequency is well beyond the receiver bandwidth, so the receiver demodulates the output and accurately reproduces the audio signal. Early reports suggested that the sound quality of class D amplifiers was comparable to or better than that of the preceding generation of low-current class A (Palmer et al. 1995) and class B amplifiers (Johnson and Killion 1994) used in analog aids.

In the author's experience, perhaps the main concern when listening to music is that the hearing aid output circuits can have substantially higher noise levels than the input circuits. Depending on the hearing aid chip, more than 15 dB of insertion gain (see Munro and Mueller, Chap. 9) may be required to amplify the input circuit noise above the output circuit noise at all frequencies (Zakis and Wise 2007). Therefore, output circuit noise may be audible during soft music or pauses with mild hearing loss and/or low gain. Although DSP algorithms can suppress input circuit noise, they cannot suppress noise in circuits that follow the DSP.

8.3.6 Receiver

The ideal receiver for music would be linear with low distortion up to the real-ear maximum output limits across the entire amplified bandwidth. In the author's experience, receivers can enter a nonlinear region (i.e., soft saturation) about 5 dB below the saturation sound pressure level (SSPL) (see Munro and Mueller, Chap. 9) (i.e., hard saturation) with a pure tone, and more headroom may be required to avoid distortion with wideband sounds. The risk of distortion increases if the selected receiver is inadequate for the hearing loss. Fitting adjustments such as increased output limits to avoid compression limiting of music peaks (Zakis and Fulton 2009) also increase the risk of receiver distortion. In general, receiver bandwidth tends to be inversely related to maximum output, which may not be a problem with a steeply sloping high-frequency hearing loss, where the typical 5–6 kHz receiver bandwidth can be preferred (see Sect. 8.5.1). When a higher bandwidth is preferred, it can be increased to around 7–8 kHz with judicious acoustic coupling and earmold design for behind-the-ear (BTE) aids (Killion and Tillman 1982). A wideband receiver unit composed of one or two receivers may be suitable for less severe hearing losses due to its relatively low SSPL.

Simulations of receivers with different bandwidths have shown that music sound quality can be reduced when the frequency response rolls off below 300 or 500 Hz depending on the stimulus, although not when the response rolls off above 7, 5, 3, or even 2 kHz (Arehart et al. 2011). While the latter may appear at odds with research on preferred amplified bandwidth (see Sect. 8.5.1) the receiver roll-off was modeled with a low-pass filter that had a shallow slope. A receiver with a spectral tilt of at least +3 dB or −4.5 dB per octave would also reduce sound quality (Arehart et al. 2011), although DSP algorithms can easily compensate for such a consistent tilt. Large spectral peaks due to acoustic resonances of the receiver, tubing, and/or ear canal can reduce music sound quality if sufficiently large (van Buuren et al. 1996; Arehart et al. 2011). While a single resonant peak (typical with in-ear receivers) is unlikely to reduce music sound quality, multiple peaks (typical with traditional BTE aids) can reduce sound quality unless their magnitude is below 12 dB (Arehart et al. 2011).

The aforementioned issues with air conduction receivers could be overcome with a recently developed tympanic contact actuator (TCA) that is part of a light-based contact hearing device. The TCA is custom-molded and floats on a layer of mineral oil on the canal side of the eardrum (Fay et al. 2013; Puria 2013). As shown in Fig. 8.3, a BTE sound processor transmits processed audio down the ear canal via infrared light, and the TCA converts the audio signal from light to a force that is applied to the eardrum. A preliminary safety, stability, and performance study showed a mean maximum equivalent output of 90 dB SPL at 0.25 kHz that smoothly increased to 110 dB SPL at 6 kHz before falling slightly at 10 kHz (Fay et al. 2013). The ear canal is left open and the low mean unaided insertion loss (Fay et al. 2013) allows unaided listening for frequencies where hearing is normal or near normal. The mean maximum gain before acoustic feedback of more than 40 dB from 2 to 5 kHz and above 50 dB at other frequencies (Fay et al. 2013) would reduce the need for acoustic feedback cancelers that have the potential to distort music (the latter is discussed in Sect. 8.5.4). It will be interesting to see how this technology performs when listening to music with a 10-kHz or greater bandwidth.

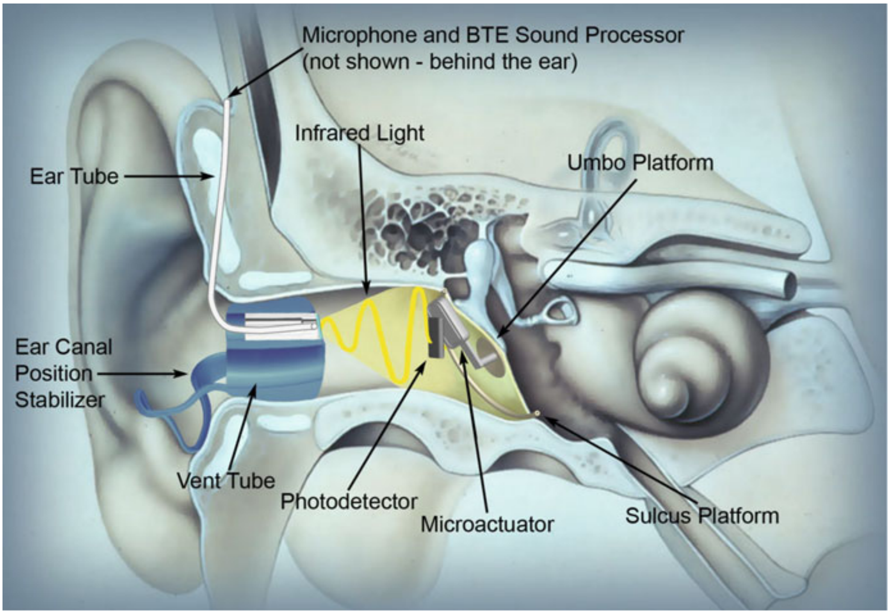

Fig. 8.3 Schematic illustration of the light-based contact hearing device, which consists of a sound processor that transmits audio via infrared light to a light-activated tympanic contact actuator (TCA), which converts the audio signal from light to a force that vibrates the eardrum. (Reproduced with permission from EarLens Corporation)

8.4 Musical Percepts

This section provides a brief overview of how well rhythm, pitch, melody, and timbre are perceived by hearing aid users. This is intended to provide context for the processing of music by DSP algorithms in Sect. 8.5. A limitation is that the literature has investigated the effects of either hearing loss (i.e., unaided) or the combination of hearing loss and hearing aids (i.e., aided) relative to normal hearing but has not isolated the effects of hearing aids from those of hearing loss (i.e., aided vs. unaided) within hearing-impaired listeners. Therefore, this overview focuses on aided listening relative to normal hearing for each musical percept.

A further limitation is that several studies used the participants' own hearing aids that differed in fitting rationale, processing, and/or method of signal presentation, and the effects of these differences were not assessed. Some studies have investigated the effects of frequency lowering on musical percepts and are discussed in Sect. 8.5.3. This overview is not intended to provide a review of the general psychoacoustics literature, which can be found in texts such as Moore (2007) and Gelfand (2010), nor the intricacies of music perception with normal hearing nor musicology.

8.4.1 Rhythm

Adult hearing aid users with moderate to severe hearing loss (Uys and van Dijk 2011) or moderately severe to profound hearing loss that meets current cochlear implant (CI) selection criteria (Looi et al. 2008a, b) perceive the rhythm of simple stimuli, such as sequences of same-pitch tones, about as well as normal. However, hearing aid users generally perform more poorly than normal on more challenging rhythm tasks that employ multiple pitches and real musical instruments (Uys and van Dijk 2011). For example, when listening to simple sequences of pure tones with the same frequency, rhythm identification based on large differences in the silent interval between tones is close to normal, but rhythm discrimination between sequence pairs that differ more subtly in tone duration and level is impaired (Uys and van Dijk 2011). Furthermore, when listening to melodies played on a piano, recognition of the rhythm as either a waltz or march is poorer than normal, as is identification of whether the first, second, or both of a pair of melodies are played in musical time (Uys and van Dijk 2011).

Processing such as heavy compression (as used in the music industry, especially for rock and pop music) can make the beat and rhythm peaks of the modulation spectrum less pronounced (Croghan et al. 2014). However, it is not clear whether such processing affects the rhythm perception of hearing aid users. More studies are required to better understand the effects of hearing aids and various processing algorithms on rhythm perception.

8.4.2 Pitch and Melody

About one in six hearing aid users have difficulty recognizing melodies in music (Leek et al. 2008). In particular, hearing aid users find it more difficult to hear individual melodic lines for rock and orchestral music (possibly owing to distortion effects and complexity, respectively) than hearing them for solo cello, solo guitar, or solo piano (Madsen and Moore 2014). The components of melody are rhythm and pitch, and the literature has tended to focus on the pitch component of melody perception.

Looi et al. (2008a, b) investigated pitch and melody perception for listeners with moderately severe to profound hearing loss meeting CI selection criteria. Pitch perception was measured through the identification of which of two successive sung-vowel stimuli (same singer and vowel) was higher in F_0. The hearing aid users performed almost as well as normal for an F_0 difference of one octave but performed progressively more poorly than normal when the difference was half and quarter of an octave. Nevertheless, pitch identification is generally better with hearing aids than with CIs (Looi et al. 2008a, b). The melody perception test used well-known melodies played on a keyboard with pitch and rhythm cues present. Closed-set melody identification ranged from about as good as normal (Looi et al. 2008a) to as poor as with a CI (Looi et al. 2008b).

Uys and van Dijk (2011) investigated pitch and melody perception for hearing aid users with moderate to severe hearing loss. The battery of tests varied in complexity and difficulty. The simple pitch perception test involved identifying whether the second of a pair of synthesized piano notes was higher or lower in F_0 than the first note. Averaged across F_0 differences that ranged from 1 to 12 semitones, identification was poorer than with normal hearing and comparable to that found in similar studies for F_0 differences of three or six semitones with greater hearing loss (Looi et al. 2008a, b). The more difficult pitch-related test was to identify whether a pair of short melodies (same rhythm) played on a piano differed in the F_0 of one or more notes, and the hearing aid users also performed more poorly than normal. Melody perception was evaluated with three tests: (1) musicality perception, in which participants indicated whether the first, second, both, or neither of a pair of melodies played on a piano sounded more musical; (2) closed-set melody identification, in which participants identified well-known songs played on a piano (with and without rhythm cues); and (3) closed-set song-in-noise identification, in which participants identified well-known songs from movie soundtracks that were mixed with car noise. The hearing aid users again performed worse than normal on all three tests.

While aided pitch and melody perception are usually worse than normal, this remains a relatively underinvestigated topic. In particular, research is needed to separate the effect of hearing aids from those of hearing loss and to evaluate the effects of DSP algorithms and bandwidth on pitch and melody perception.

8.4.3 Timbre

Timbre is the attribute of auditory perception that allows two sounds to be judged as dissimilar using any criterion other than pitch, loudness, or duration (ANSI 1960). Timbre depends on a combination of sound features, especially the temporal envelope, temporal onset characteristics, spectral centroid, and spectral fine detail (Plomp 1970; Grey 1977; Caclin et al. 2005). Since timbre allows two musical instruments that play the same note at the same loudness to be distinguished, aided timbre perception is usually evaluated through the ability to identify different instruments.

Musical instrument and ensemble identification was investigated using listeners with moderately severe to profound hearing loss meeting CI selection criteria (Looi et al. 2008a, b). Musical instrument identification was evaluated with 12 instruments that covered a wide range of F_0s and four instrumental families. The instrument sounds were presented in isolation for one test and with background orchestra accompaniment for another test. The ensemble identification test used 12 ensembles that covered a range of genres and numbers of instruments. Closed-set instrument and ensemble identification was poorer than normal for the hearing aid users in all tests and was worse for the multiple-instruments test than for the single-instrument test. The latter finding is consistent with another study in which hearing aid users with such hearing loss found it less pleasant to listen to multiple instruments than to a solo instrument (Looi et al. 2007). Although such aid users generally had pitch perception superior to that of CI users, their timbre perception was often as poor as that of CI users (Looi et al. 2008a, b).

A different set of tests was used to investigate musical instrument identification for hearing aid users with moderate to severe hearing loss (Uys and van Dijk 2011). Single- and multiple-instrument identification was evaluated with eight musical instruments that represented different F_0 ranges and four instrumental families. Each instrument played a melody in isolation for the single-instrument identification task, while two or three instruments played the same melody in unison (stereo sound field) for the multiple-instrument identification task. In both cases, closed-set identification of the instrument(s) was poorer than normal. However, the ability to detect how many different instruments played in short pieces of music was almost as good as normal. This may have been a relatively easy task because of the use of five instruments with distinctly different timbres.

In summary, aided timbre perception, assessed via instrument and ensemble identification, is generally worse than normal for hearing aid users. However, more research is required to separate the effects of hearing aids from hearing loss. Studies could investigate the effects of various hearing aid algorithms on the temporal and spectral cues responsible for timbre perception and the required processing to achieve the most realistic timbre perception with musical instruments.

8.5 Digital Signal Processing

Only 40 % of hearing aid users report having a special music program, and in general these programs do not lead to improved loudness, clarity, distortion, or tonal balance but can lead to a small improvement in the ability to hear individual musical instruments (Madsen and Moore 2014). The small to nonexistent benefit may be because it is far from clear what combinations of DSP algorithms and settings should be used in music programs. In addition, as discussed in Sect. 8.3.4, the clock speed of hearing aid DSPs is limited to roughly 1 % of that of a current personal computer to achieve a good battery life. This limits the number and complexity of algorithms that can be implemented, which is reduced further if the sampling rate is increased to extend the music bandwidth. Although it may be desirable to run fewer algorithms in a music program, more computationally intensive algorithms (e.g., distortion-free acoustic feedback canceler) may be required for improved music perception. A few studies have investigated the preferred amplification settings, bandwidth, and number of frequency channels for music, whereas fewer studies have investigated the effects of other algorithms. Most studies do not delve into the effects of different algorithms and settings on rhythm, timbre, pitch, and/or melody perception. The remainder of this section discusses the literature on the use and configuration of different types of DSP algorithms for listening to music. The functionality of the DSP algorithms is described by Launer, Zakis, and Moore, Chap. 4.

8.5.1 Amplification and Bandwidth

Prescriptive procedures for generating WDRC fittings from the audiogram, such as NAL-NL1 (Byrne et al. 2001), NAL-NL2 (Keidser et al. 2011), DSL 5.0 (Scollie et al. 2005), and CAM2 (Moore et al. 2010), are tailored for a speech input and generally aim to amplify speech so that it is audible, intelligible, and not too loud. It seems unlikely that prescribed gains, compression ratios, and/or compression thresholds that were shaped across frequency based on the spectrum, dynamic range, loudness, and/or intelligibility of speech will be optimal for listening to music. This may explain why hearing aids can make music seem lacking in bass, too bright/shrill, have worse tone quality, and make loud parts too loud or soft parts too soft (Madsen and Moore 2014). An alternative of simply aiming to fit WDRC to fully restore normal loudness perception for all sounds over the normal range of hearing is neither practical, owing to the high gains required at low input levels, nor preferred by hearing aid users (Smeds 2004; Smeds et al. 2006). The ADRO algorithm has the opposite approach, aiming to make the most important parts of the signal audible and comfortable over a wide range of input levels using slow time constants (Blamey 2005). Given that a typical music spectrum cannot be specified, one possible fitting approach would be for the hearing aid user to train the aid's amplification parameters to his or her preference (Zakis et al. 2007) in a dedicated music program while listening to his or her favorite music.

Another issue is that while music can have significant energy for frequencies up to and above 10 kHz, some WDRC prescriptive procedures such as DSL 5.0, NAL-NL1, and NAL-NL2 only prescribe settings for frequencies up to 6, 6, and 8 kHz, respectively. Similarly, ADRO's standard fitting prescriptions extend only up to 6 kHz. An exception is CAM2, which prescribes WDRC settings up to 10 kHz and has recently been refined and evaluated with a simulated hearing aid (Moore et al. 2011; Moore and Sęk 2013).

As a consequence of the general speech bias of prescriptive procedures, hearing aid manufacturers have modified the prescribed and/or nonprescribed (e.g., time constants) amplification parameters for use in specialized music programs. Due to a lack of research literature to guide this process, manufacturers have taken the lead and independently developed different and sometimes opposing fitting approaches for music programs (Chasin and Russo 2004). It is not clear what approach is best and whether this depends on hearing loss. Some degree of difference in approach across manufacturers may be required due to differences in the characteristics of their underlying signal path filtering algorithms, signal-processing architectures, amplification algorithm implementations, and interactions with other algorithms. However, the adjustment of amplification parameters due to such differences is largely up to manufacturers, while the literature has generally used simulated aids with a generic signal path running only the amplification algorithm. The remainder of this subsection discusses the literature on preferred amplification settings and bandwidth for music.

8.5.1.1 Bandwidth

Extension of the aided bandwidth beyond 5–6 kHz has the potential to improve music sound quality and perceived timbre. Studies with simulated hearing aids allow investigations into the possible benefits of amplifying frequency components above 5–6 kHz that are not limited by current hearing aid technology. Such studies have shown that normal-hearing listeners prefer an upper cutoff frequency of 9 kHz over 5.5 kHz when listening to music with WDRC processing (Ricketts et al. 2008), which is consistent with previous research with linear processing (Moore and Tan 2003). Mild to moderately impaired listeners also generally prefer a cutoff frequency of 9 kHz over 5.5 kHz (Ricketts et al. 2008), while listeners with greater mean high-frequency hearing loss have no mean preference among cutoff frequencies of 5, 7.5, or 10 kHz (Moore et al. 2011), and listeners with mild to severe hearing loss have no mean preference between cutoff frequencies of 5 or 11 kHz (Brennan et al. 2014). These studies are more consistent when individual data are considered because preferences for a higher cutoff frequency were associated with shallower high-frequency audiogram slopes and preferences for a lower cutoff frequency were associated with steeper high-frequency audiogram slopes (Ricketts et al. 2008; Moore et al. 2011). Furthermore, preferences for a lower or higher cutoff frequency have been associated with relatively poor or good high-frequency (4–8 kHz) hearing thresholds, respectively (Brennan et al. 2014), and a lower cutoff frequency tends to be preferred when hearing loss exceeds 70 dB at 12 kHz (Ricketts et al. 2008). Preferences may depend on the high-frequency energy of the stimulus (Moore et al. 2011) and were independent of WDRC time constants (Moore et al. 2011) and of two signal-processing architectures that were evaluated (Ricketts et al. 2008).

A major difference across studies was the use of different prescriptions: CAM2, which prescribes in a consistent way up to 10 kHz (Moore et al. 2011), NAL-NL1 up to its 6-kHz prescription limit with settings at 8, 10, and 12 kHz that were acknowledged as somewhat arbitrarily derived (Ricketts et al. 2008), or DSL 5.0a up to its 6-kHz prescription limit with 8-kHz settings based on the 6-kHz values (Brennan et al. 2014). Therefore, while the evidence suggests that preferred upper cutoff frequency is related to high-frequency hearing loss, this could be investigated in more detail with regard to the effects of hearing loss, prescriptive procedure, music stimulus, and input level.

Apart from hearing loss, preferences for amplification of frequencies above 5–6 kHz are affected by the relative high-frequency gain. Preferences between CAM2 and CAM2 with reduced (10–20 % less in dB) or increased (11–25 % more in dB to 50 dB limit) gain from 4 to 10 kHz have been investigated with a simulated WDRC aid using fast (attack/release = 10/100 ms) or slow (50/3,000 ms) time constants (Moore et al. 2011). Regardless of time constants and type of music stimuli, mild to moderately impaired participants preferred the CAM2 and reduced high-frequency gains about equally but did not prefer the increased high-frequency gains. The similar preference between gains that substantially differ suggests that the "optimal" gains, on average, may lie between the CAM2 and reduced high-frequency gains. Another possible explanation is that most participants were inexperienced aid users who were fitted with an early version of CAM2 that did not prescribe less gain for such users. Participants who tended to prefer the CAM2 over reduced high-frequency gains also tended to prefer the 7.5-kHz cutoff frequency over 5 kHz in another experiment, averaged across speech and music stimuli (Moore et al. 2011). So although it appears that some people have either a clear like or dislike of high-frequency amplification, the reasons for this could be further investigated with a wider range of music stimuli at levels other than 65 dB SPL.

The bass end of the aided bandwidth has received less attention than the treble end. It has been argued that amplifying below approximately 125 Hz to include the F_0 of instruments such as the bass or cello may be unnecessary, as the F_0 is not required to perceive pitch and the mid- to high-frequency harmonic structure typically defines perceived quality (Chasin 2012). In addition, not amplifying below 125 Hz could be advantageous in noisy environments, especially if noise reduction algorithms are turned off for music (Chasin 2012). On the other hand, it has also been argued that audibility of the fundamental component is important for hearing the balance of one note against the other and for aspects of timbre such as the natural warmth and fullness of low-F_0 notes (Revit 2009). Normal-hearing listeners perceive music as more natural as the lower limit of the audio bandwidth is progressively decreased from 313 to 55 Hz (Moore and Tan 2003). Early work with hearing-impaired listeners and one music stimulus also showed that progressively lower cutoff frequencies (650, 200, and 90 Hz) were preferred, although this was done with an analog aid that had no flexibility to finely shape the gain across frequency to compensate for the hearing loss or receiver response (Franks 1982). More recent simulations of receiver roll-off via high-pass filtering have shown that with NAL-R prescribed linear gain (Byrne and Dillon 1986), a cutoff frequency no greater than 200 or 300 Hz is required to avoid reduced music sound quality depending on the stimulus (Arehart et al. 2011). It should be noted that a low-order filter was used (i.e., gradual roll-off) so harmonics below the cutoff frequency may have remained audible (but at lower levels). Preferences for the low cutoff frequency of amplification and the effect on pitch and/or timbre perception could be investigated further with a greater range of hearing losses, fitting prescriptions, and music stimuli.

8.5.1.2 WDRC Time Constants

Compression time constants are often referred to as “fast” or “slow,” but these terms can be applied to a wide range of compressor speeds and may not equally apply to the attack and release times. For clarity, the attack/release times used in studies will be given where possible. Time constants may potentially affect aspects of timbre (e.g., onsets) and loudness contrasts. The perceptual effects of WDRC time constants are usually evaluated with linear amplification as the sound quality benchmark owing to its low temporal distortion. Compared with linear processing with NAL-R gain, the addition of temporal distortion from simulated 18-channel, fast-acting WDRC (5/70 or 5/200 ms in all channels) did not significantly affect the mean sound quality ratings of listeners with mild to moderately severe hearing loss for classical, jazz, or vocal music at 72 dB SPL (Arehart et al. 2011). The compression ratio (CR) was not set for the hearing loss (CR = 2) so all participants received the same amount of temporal distortion. A more extreme CR of 10 with time constants of 5/10 or 5/70 ms reduced rated sound quality for classical and jazz music but not for vocals (Arehart et al. 2011). However, this combination of CR and time constants is more typical of compression limiting than of WDRC amplification.

Although linear amplification and WDRC may receive equivalent sound quality ratings, they may not be equally preferred with more sensitive paired-comparison methodologies. The latter have shown that when linear gain (NAL-R) and WDRC (NAL-NL1) fittings in a simulated hearing aid were adjusted to give similar loudness for stimuli at 65 dB SPL, linear gain and slow WDRC (50/1,000 ms) were equally preferred to fast WDRC (5/50 ms) for classical music, while linear gain was preferred to slow WDRC and slow WDRC was preferred to fast WDRC for rock music (Croghan et al. 2014). Apart from the stimulus, preferences for WDRC time constants can also depend on the music input level. This was investigated with a simulated aid and CAM2 fitting (Moore et al. 2011). Slow time constants (50/3,000 ms) were preferred to medium (20/300 ms) and fast (10/100 ms) time constants for jazz and classical music at 80 dB SPL, while slow time constants were only preferred to fast for classical music at 65 dB SPL. The lack of any significant preference for the 50 dB SPL input level may have been due to the stimuli dipping into a linear region below the compression thresholds (CTs).

Overall, the research has primarily investigated the effect of WDRC release times because the attack times were generally much faster and varied over a smaller range. The results show a general preference for slower release times with music. However, it is not clear what has driven these preferences. Possible factors include timbre, loudness dynamics, temporal distortion, and perhaps the lower average gain over time when the release time is much slower than the attack time (if loudness was in fact greater than preferred for higher input levels). Another possible factor is how clearly individual instruments and vocals are heard in pieces of music. This was investigated with a simulated 5-channel aid, CAM2 fittings, and classical, jazz, and pop music at 70 dB SPL (Madsen et al. 2015). Clarity was greater for linear amplification than for WDRC, and some participants consistently rated clarity higher for slow (50/3,000 ms) than fast (10/100 ms) WDRC. The latter may be partly due to greater temporal envelope distortion from fast than slow WDRC. Clarity was not reduced by cross-modulation (i.e., the partial correlation of the envelopes of different sounds in the same channel produced by the time-varying gain), possibly because the envelopes of instruments playing in the same piece of music can be inherently well correlated.

8.5.1.3 WDRC Channels and Fittings

It is surprising that few studies have compared the efficacy of different WDRC prescriptions that could differ in how they affect timbre and loudness dynamics. Sound quality preferences for CAM2 and NAL-NL2 have recently been compared with a simulated aid (Moore and Sęk 2013). As shown in Fig. 8.4, these prescriptions generally differed in gain by less than 5 dB for frequencies below 4 kHz, while CAM2 prescribed progressively more gain than NAL-NL2 from 4 to 10 kHz (because NAL-NL2 prescribes only for frequencies up to 8 kHz, its 8-kHz settings were used at 10 kHz). Participants with mild hearing loss up to 4 kHz preferred CAM2 to NAL-NL2 regardless of stimulus (jazz trio, orchestra, xylophone, or countertenor accompanied by guitar and recorder), input level (50, 65, or 80 dB SPL), compressor speed (10/100 or 50/3,000 ms), or degree of high-frequency hearing loss. However, mean preferences were small (less than "slightly better") and smallest at the more typical level of live music (80 dB SPL). It is not clear why CAM2 was preferred (e.g., timbre, loudness, dynamics, etc.), so it is difficult to predict how preferences may change with real hearing aids if the CAM2 high-frequency gains need to be reduced to avoid acoustic feedback. The small differences in mean preference may have been due to both prescriptions being primarily designed for speech inputs, while other approaches may be preferable for music.

Clinicians have been modifying WDRC prescriptions to improve music sound quality for decades (Chasin 2006). Clinical recommendations have included using just one WDRC channel or, alternatively, multiple WDRC channels with similar CTs and CRs across frequency, with the aim of achieving a good balance between the F_0 and its harmonics (Chasin 2006). The use of similar CTs and CRs across frequency is similar to a loudness normalization approach for a listener with a flat

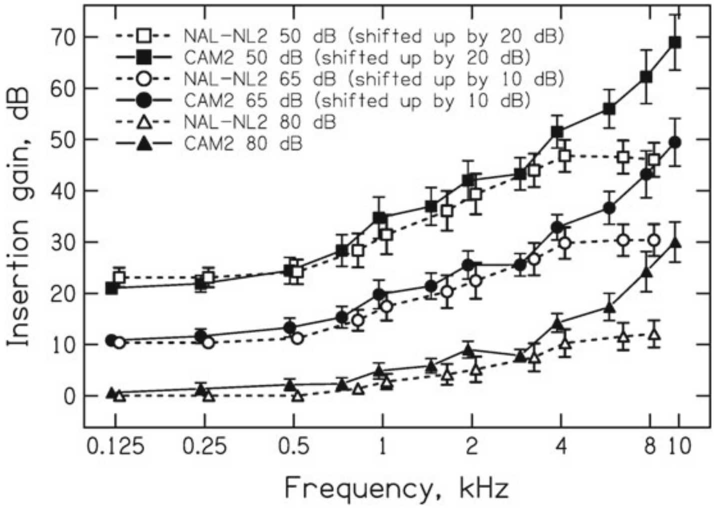

Fig. 8.4 Mean insertion gains prescribed by NAL-NL2 and CAM2 across participants for speech-shaped noise at 50, 65, and 80 dB SPL. The values for 65 and 50 dB SPL were shifted up by 10 and 20 dB, respectively, for clarity [Adapted from Moore and Sęk (2013)]

audiogram and similar loudness discomfort levels across frequency. For a listener with a dynamic range of hearing that varies greatly with frequency, some shaping of the CR across frequency may be required to avoid inaudibility or loudness/limiting issues, while with one WDRC channel a strong bass sound could drive down the gain enough to make simultaneous high-frequency sounds inaudible.

Few studies have investigated the efficacy of adjustments to CTs, CRs, and/or channel numbers on music perception. Acoustically, more compression channels and faster time constants reduce the wideband dynamic range of music (Croghan et al. 2014). Compared with linear amplification, compression with the CT set 30 dB below the stimulus root-mean-square (RMS) level, the same CR in all channels, and the fast time constants of a temporal envelope low-pass filter (32 Hz cutoff) generally reduced the sound quality of classical and rock music with increasing CR (1, 2, or 4) and number of channels (1, 4, or 16), although sound quality was equivalent for some conditions (van Buuren et al. 1999). For a more typical case of gains and CRs prescribed by NAL-NL1, fewer channels (3 vs. 18) were preferred with rock music and more so with faster WDRC time constants (5/50 ms) or heavier compression limiting (as used in commercial music recordings) applied to the stimulus (Croghan et al. 2014). However, 3 or 18 channels were equally preferred with classical music, on average, with slight individual preferences for fewer channels associated with worse frequency selectivity at 0.5 kHz but not at 2 kHz (Croghan et al. 2014). Although these studies suggest that fewer WDRC channels and lower CRs can be preferable for music, this depends on the time constants, fitting method, stimulus, and hearing loss. It is not clear how these factors affected percepts such as timbre, pitch, loudness dynamics, and so forth and how such percepts determined overall preferences.

8.5.1.4 Adaptive Dynamic Range Optimization

Instead of WDRC, some hearing aids use ADRO, which aims to place the most important part of the input dynamic range within an audible and comfortable range of output levels while preserving short-term level contrasts and the integrity of the temporal envelope in each channel (Blamey 2005). This is accomplished by adjusting the gain very slowly (typically by 3 dB per second) or not at all, based on a statistical analysis of the recent output signal and the sequential application of four processing rules independently in each channel (ADRO aids typically have 32 or 64 linearly spaced channels). Figure 8.5 shows a 32-channel ADRO prescription for a flat 50-dB hearing loss. The ADRO fitting parameters are derived from the audiogram and comfortable output levels, and the latter can be predicted from the audiogram or individually measured with noise bands presented by the aid in situ. The maximum output limits (MOLs) are the thresholds of fast-acting compressive output limiters. The comfort targets determine the upper limit of the desired range of output levels, and the audibility targets determine the lower limit of this range for sounds of interest (within the constraint of the maximum gains, which define the upper gain limit).

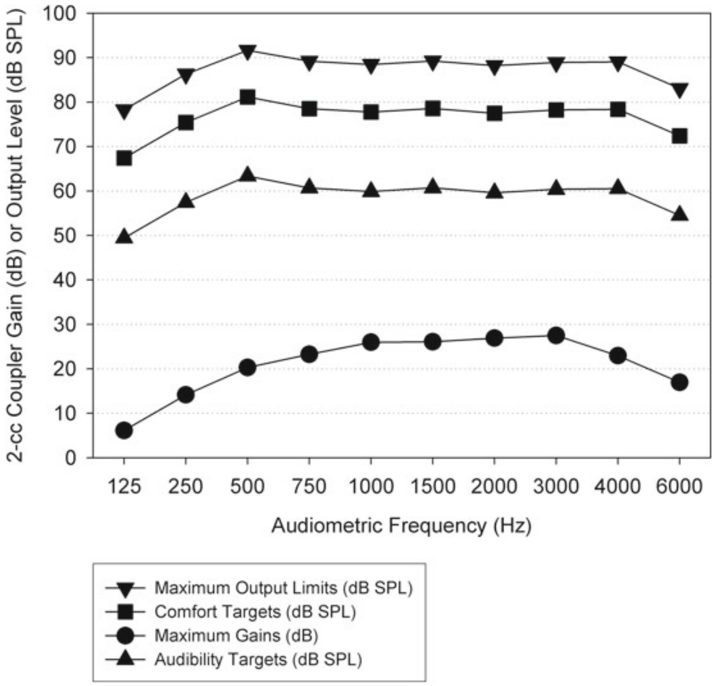

Fig. 8.5 An ADRO fitting prescription for a flat 50-dB hearing loss. The maximum output limits (*inverted triangles*), comfort targets (*squares*), and audibility targets (*upright triangles*) are in 2-cc coupler output level units, while the maximum gains (*circles*) are in 2-cc coupler gain units

The first ADRO rule applies the MOLs to protect against excessive loudness. The second rule causes ADRO to slowly reduce the gain if the output level that is exceeded 10 % of the time has risen above the comfort target to maintain loudness comfort. If the gain was not reduced due to the second rule, the third rule causes ADRO to slowly increase the gain if the output level that is exceeded 70 % of the time has fallen below the audibility target to improve the audibility of soft sounds. The fourth rule limits the gain to the maximum gain, so that soft sounds are perceived as soft and lower-level input circuit noise is not overamplified. In a nutshell, ADRO aims to keep at least the upper part of the recent dynamic range of signals of interest audible and comfortable, using slow time constants to minimize temporal envelope distortion.

The efficacy of ADRO and WDRC hearing aids for music was compared with receiver-in-the-ear devices and each manufacturer's default program (not a music program) and first fit settings (Higgins et al. 2012). The ADRO aids had 32 channels, while the WDRC aids used dual fast (4-channel) and slow (15-channel) processing. In both systems, only the acoustic feedback management systems were turned on (directional microphones and noise reduction were disabled). Sound quality ratings were higher with the ADRO aids for all three music stimuli (classical, jazz, and rock at 70 dB SPL). This was not at the cost of speech understanding in noise, which was also better with the ADRO aids. Analysis of the music output signals (recorded with an artificial ear) showed that the ADRO aids provided greater output around 1.5 kHz and a generally lower output at other frequencies, and the WDRC aids compressed the short-term dynamics of the music. This was consistent with participants' comments that the WDRC aids sounded warmer and had more bass, while the ADRO aids sounded brighter, clearer, and crisper and had greater definition, and that it was the brightness and clarity that dominated overall preference. It is interesting to compare this study, where a 32-channel ADRO aid was preferred to a 4/15-channel WDRC aid, to other studies where increasing numbers of channels were *less* preferred within WDRC aids and especially with faster time constants. It is not clear whether the aid with more channels was preferred because of its ADRO processing, the very slow time constants, differences in fitting goals between aids, or some other factor(s).

While music sound quality was highly rated with ADRO prescriptions generated from the audiogram (Higgins et al. 2012), clinical experience has shown that very particular clients such as musicians can appreciate a fine-tuned music program (Zakis and Fulton 2009). Clinical adjustments have included flattening the maximum gain across frequency to improve tonal balance, increasing the maximum gain from 0 to 1 kHz to give a fuller sound quality, and increasing ADRO's relatively conservative MOLs to be above the most intense music peak levels (Zakis and Fulton 2009).

8.5.2 Signal Path Delay and Phase

The effects of the delay hearing aids impose on the audio signal have been extensively investigated with speech stimuli. In comparison, only two studies have investigated delay effects with music. Both studies used ear-level, open-canal devices but investigated different delay effects with different methodologies. In the first study (Groth and Søndergaard 2004), mild to moderately impaired participants were fitted with BTE casings connected to a digital equalizer, which provided a flat, linear insertion gain of 10 dB from 1 to 6 kHz (steeply rolled off below 1 kHz) and electronic delays of 2, 4, and 10 ms. All delays were rated as “not at all disturbing” for acoustic instruments presented at a comfortable loudness. Owing to the open fitting and steep gain transition, the dominant delay percept was probably the temporal desynchronization between the unaided and aided (delayed) sound that dominated below and above 1 kHz, respectively. The strength of this percept may depend on the spectrotemporal properties of the music, such as whether an instrument produces sound simultaneously at unaided and aided frequencies, or the steepness of temporal onsets. Such potential factors should be investigated before maximum acceptable temporal desynchronization limits for music can be defined.

In the second study (Zakis et al. 2012), mild to moderately hearing-impaired musicians were fitted with real hearing aids and a linear gain that was individually adjusted to maximize spectral ripple depth (owing to aided and unaided sounds interacting in the ear canal) from 0.75 to 6 kHz. Delay and the signal-path FIR filter’s phase-frequency response were varied to give three aided processing conditions that differed in the locations of the spectral ripples and had nominal acoustic delays of 1.4 ms (minimum phase), 3.4 ms (linear phase), and 3.4 ms (minimum phase). An unaided (muted output) condition served as the sound quality benchmark. Although individual preferences could be strong, there was no median preference difference between aided conditions or between unaided and aided conditions when listening to two classical music stimuli presented at mezzo forte loudness. This study also showed that hearing aid circuits can apply a much greater variation in phase or delay across frequency than the evaluated FIR filter designs. Further research with more music stimuli, longer delays, other types of digital filtering, and different gains is required before general conclusions can be drawn about the acceptance of spectral ripples when listening to music with open-canal aids.

8.5.3 Frequency Lowering

Many frequency-lowering approaches have been devised over several decades to improve the audibility of high-frequency information in speech (Simpson 2009). The effects of such processing on the perception of music have received little attention. This class of algorithms has the potential to increase the audibility of high-frequency components in music, which would otherwise be inaudible due to the severity of the hearing loss, perceived as distorted due to cochlear dead regions, or simply beyond the bandwidth of a power receiver. Users of frequency-lowering aids report less acoustic feedback while listening to music compared with users of other devices (Madsen and Moore 2014). Differences in how these algorithms work across manufacturers could influence the audible bandwidth, the perception of pitch and timbre, and music enjoyment in general. However, only the efficacy of the linear frequency transposition (LFT) and nonlinear frequency compression (NLFC) algorithms has been investigated with music, so only these algorithms are considered in the text that follows.

The LFT algorithm continuously identifies the most prominent spectral peak at high frequencies, and a one-octave-wide band around that peak (the source octave) is shifted down by one octave, bandpass filtered (to the destination octave width), amplified, and mixed with the original signal (Kuk et al. 2009). Higher harmonics can be transposed on top of and/or between lower harmonics (McDermott 2011), which could potentially affect timbre and pitch, although pitch effects are unlikely because even after transposition the harmonics are not resolved. Children with a severe to profound high-frequency hearing loss (candidates for LFT) found LFT to be as preferable as or more preferable than no LFT about 60 % of the time for 10 music samples (Auriemmo et al. 2009). Adults with a moderately severe high-frequency hearing loss (less clear candidates for LFT) with fittings based on NAL-NL1 found LFT, no LFT with limited bandwidth (upper cutoff frequency 4 kHz), and no LFT with extended bandwidth (upper cutoff frequency 8.35 kHz) to be equally preferable and have similar degrees of naturalness, distortion, and loudness for five music samples (Lau et al. 2014). However, in real-life conditions LFT was clearly the least preferred option when listening to music (Lau et al. 2014). The reasons behind this preference are not clear and warrant further investigation.

The NLFC algorithm reduces the frequency of components above the cutoff frequency by an amount that increases with increasing frequency. The source frequency range divided by the destination frequency range (both expressed in octaves) is the frequency compression ratio (FCR). Compressed frequencies remain above the cutoff frequency. NLFC processing reduces the spacing between harmonics above the cutoff frequency, introducing inharmonicity that may potentially affect timbre, but is unlikely to affect pitch because resolved harmonics are likely to fall below the cutoff frequency (McDermott 2011). When used with DSL 5.0 fittings and nonindividualized NLFC parameters (FCR of 2), NLFC may or may not reduce the sound quality of music depending on the stimulus (pop or classical music) and NLFC cutoff frequency (Parsa et al. 2013). With prescribed and where necessary fine-tuned NLFC settings, moderately to severely hearing-impaired participants could hear a slightly greater number of musical instruments with very different timbres in short musical pieces with than without NLFC, while NLFC did not affect the identification of single and multiple simultaneous instruments, well-known melodies and songs, or the perception of whether melodies sound musical (Uys et al. 2013). In real-life situations, NLFC reduced tininess and reverberance and improved the perception of rhythm and the detection of different instruments in a musical piece (Uys et al. 2012). Rated loudness, fullness, crispness, naturalness, overall fidelity, pleasantness, ability to hear lyrics and melody, and ability to distinguish between high and low notes were not significantly affected (Uys et al. 2012).

Although frequency-lowering aids are generally not associated with increased reports of distortion (Madsen and Moore 2014), research could be directed toward more sensitive evaluations of the effects of different algorithms on the perception of timbre and pitch for different instruments, musical notes, types of aid users (e.g., musicians vs. nonmusicians), and degrees of hearing loss. The latter has become increasingly important as the fitting range of some algorithms has been extended to less severe hearing losses. Whether frequency-lowering algorithms should be used when listening to music may ultimately depend on the individual benefits for audibility and acoustic feedback reduction versus the perception of spectral distortions.

8.5.4 Acoustic Feedback Cancelation and Suppression

Hearing aids use a wide range of acoustic feedback cancellation and/or suppression algorithms that can misclassify tonal sounds in music as acoustic feedback. As a result, suppression algorithms can reduce the gain around harmonic frequencies, and cancellation algorithms can adapt to the music stimulus instead of the feedback path, which results in the cancellation of music harmonics and the addition of extraneous spectral components (Freed and Soli 2006). To reduce these problems, some acoustic feedback cancelers have a music mode in which adaptation to the feedback path is slowed, constrained, or even stopped (Freed and Soli 2006; Spriet et al. 2010). Although this reduces the chance of distortion from the algorithm following misidentification of tonal sounds as feedback, it limits acoustic feedback control when the feedback path changes. Even with a constant feedback path, feedback control can be lower with opera music than with speech when the feedback canceler is operating in music mode or normal mode (Spriet et al. 2010). Other approaches for avoiding adaptation to music include decorrelating the input and output signals by introducing a frequency or phase shift, although this has the potential to affect pitch and timbre (Freed and Soli 2006; Spriet et al. 2010) and can result in “beating” effects with open-fit hearing aids. More than one-third of aid users report acoustic feedback while listening to music (Madsen and Moore 2014), and clinical reports have verified that acoustic feedback reduction systems can react to highly tonal instruments such as the flute (Chasin and Russo 2004). This has led to recommendations that feedback reduction systems be turned off in music programs if not needed (Chasin and Hockley 2014).

Few studies have investigated the perceptual effects of feedback reduction algorithms used in commercial hearing aids when listening to music. In one study (Johnson et al. 2007), hearing aids with two different frequency-domain cancellation systems were fitted to a participant with mild to moderate hearing loss, and the outputs were recorded for a flute concerto excerpt presented at 50 dBA. The recorded signals were given frequency-dependent amplification appropriate for the participant with the worst hearing loss and presented to all participants. Neither a constrained adaptation system nor a slow adaptation system with adaptive notch filtering significantly affected overall sound quality.

In another study (Zakis et al. 2010), mild to moderately hearing-impaired musicians were fitted with open-canal aids and a flat, linear insertion gain that aimed to match the levels of the aided and unaided sounds in the canal from 0.75 to 6 kHz. Such a fitting can be problematic for accurate adaptation to the feedback path, as the fed-back sound can be dominated by the input stimulus at the microphone. There was no significant difference in overall preference or artifact perception depending on whether a fast-adapting, nonconstrained, time-domain feedback canceler was on or off during a passage of jazz music that was known to be problematic for simpler feedback cancelers.

Although these findings may seem to contradict clinical reports, they were based on a relatively small sample of hearing aids, algorithms, fittings, and music stimuli. In the author's experience, feedback cancelers certainly do have the potential to produce strong artifacts with music, and the occurrence, type, and severity of the artifacts depend on factors such as the algorithm design, interactions with other algorithms, fitting, and stimulus type and level.

8.5.5 Directional Microphones

A directional microphone may be beneficial when listening to music in some situations. For example, when listening to live music, a fixed directional microphone may help place the focus on the stage rather than the disruptive activities of the audience (Hockley et al. 2012). When the forward gain-frequency response is matched for omni- and fixed-directional microphone patterns (cardioid, supercardioid, and hypercardioid), the sound quality of classical music presented in a diffuse sound field at 75 dBA was similar across microphone types for all evaluated dimensions (fullness, nearness, brightness, fidelity, spaciousness, loudness, softness, and clarity) (Bentler et al. 2004). It has been suggested that adaptive directional microphones should be turned off because they may treat music as noise (Hockley et al. 2012). Further studies could quantify the trade-offs between the use of fixed and adaptive directionality in different spatial listening situations and investigate whether the increased low-frequency circuit noise of directional microphones is problematic when listening to very soft music or during pauses.

8.5.6 Noise Reduction

This section discusses the use of algorithms that target five different categories of sound that are treated as unwanted sounds: input circuit noise, background noise, wind noise, transient sounds, and reverberation.

8.5.6.1 Input Circuit Noise Reduction

Noise generated by input circuits such as microphones, preamplifiers, and ADCs has the potential to be audible with sufficiently mild hearing loss or high gain. Low-level expansion in hearing aids suppresses this noise, and potentially also low-level background noise, by reducing the gain when the input level falls below the expansion threshold. The amount of gain reduction is determined by the expansion ratio. The literature has focused on the effects of expansion on speech understanding and low-level noise but not music perception. Expansion could well suppress soft music, as some expansion settings can reduce speech understanding in quiet or even noise (Plyler et al. 2005, 2007; Zakis and Wise 2007). However, in everyday situations, expansion is preferred to no expansion in quiet or low-level background noise (Plyler et al. 2005, 2007). The use of microphones with a sensitivity that rolls off by 6 dB/octave below 1 kHz to avoid clipping at music peak levels (Schmidt 2012) or a fixed directional response to put more focus on the stage (Hockley et al. 2012) results in decreased low-frequency sensitivity and hence decreased signal-to-noise ratio in the microphone circuits. Increased DSP gain to compensate for the reduced sensitivity also increases the circuit noise level and hence the need for expansion that may also suppress soft music. However, it remains unclear whether expansion causes audibility or distortion issues with soft music and whether unsuppressed circuit noise is preferable to a small reduction in audibility.

8.5.6.2 Background Noise Reduction

This class of signal-processing algorithm generally aims to preserve the level of sounds considered to be speech and reduce the gain for sounds considered to be noise. Many algorithms tend to identify sounds with high modulation depths and/or speechlike modulation rates (around 4 Hz) as desired and sounds with lower modulation depths and different modulation rates as unwanted (Bentler and Chiou 2006). As discussed in Sect. 8.2, music can have similar or very different temporal characteristics to speech, so some instruments can be treated more like noise than others. This results in the suppression of instruments such as the guitar, saxophone, and piano by differing amounts across frequency within and across algorithms (Bentler and Chiou 2006). In addition, algorithms that quickly vary the gain with the modulation depth have the potential to desynchronize the temporal envelope across different channels, which may impair source segregation and possibly pitch and timbre perception. In light of the preceding, it is not surprising that recommendations have been made to turn off background noise reduction wherever possible to avoid the distortion of music (Chasin 2012; Chasin and Hockley 2014).

Some hearing aids analyze modulation across channels to determine whether a sound with a harmonic structure is present (wanted signal) or absent (unwanted noise). This approach largely avoids the suppression of the guitar, saxophone (with background accompaniment), and piano (Bentler and Chiou 2006).

However, it is not clear how well this works with multiple instruments (where the modulation may differ in each channel due to different combinations of harmonics from different instruments), nonharmonic musical sounds, or harmonic noises (e.g., from motors). Further research could investigate the efficacy of different algorithms for reducing noise when listening to music or whether there is in fact a need for such algorithms.

8.5.6.3 Wind Noise Reduction

Wind noise is due to turbulence in air that flows across the microphone port. Wind noise increases in level and extends to higher frequencies with increasing wind speed and varies with wind angle, microphone location, and across and within styles of hearing aid (Chung et al. 2009, 2010; Zakis and Hawkins 2015). There is great potential for wind noise to mask music at outdoor events because wind noise levels can be as high as 94 and 109 dB SPL at wind speeds of just 3 m/s and 6 m/s, respectively (Zakis and Hawkins 2015). Wind noise levels are limited to around 110–115 dB SPL at 12 m/s due to microphone saturation (Zakis 2011). Positive preamplifier gain could cause saturation in preamplifier circuits, and hence distortion of music in the presence of the wind noise, at much lower wind speeds than 12 m/s.

There is substantial potential for music to be distorted by wind noise reduction algorithms in two ways: (1) incorrect classification of musical sounds as wind, which may lead to the application of large gain reductions in the absence of wind; and (2) the application of large gain reductions to suppress wind noise that is present, which inadvertently distorts and/or suppresses music. No studies have investigated whether wind noise detection (WND) algorithms correctly classify music as not wind, although one study has shown that WND algorithms differ in their accuracy in classifying pure tones (Zakis and Tan 2014). Music may potentially be treated as a pure tone if each harmonic (i.e., tone) is in a different channel and WND is performed independently in each channel. Most WND algorithms assume that wind is present when the outputs of the two microphones differ sufficiently in level and/or phase (indicating localized turbulence) and that wind is not present when the microphone outputs are similar (indicating a far-field propagating wave). However, microphone outputs in response to pure tones can be quite dissimilar in level and/or phase due to acoustic reflections from the room, head, and/or pinna (Zakis and Tan 2014).

Owing to the high levels of wind noise, rather large gain reductions can be applied when wind noise is detected (based on manufacturers' white papers, up to 40 dB at low frequencies and less at high frequencies). Such gain reductions could severely distort the music spectrum if incorrectly applied in the absence of wind. In the presence of wind, they would ideally reduce the loudness of wind noise in channels where it dominates music, which may also reduce the masking of music by wind at other frequencies. However, if less appropriately applied, such gain reductions could suppress and distort music at frequencies where wind noise does not dominate. The effects of wind noise reduction algorithms when listening to music outdoors (whether positive or negative) require further investigation. Such algorithms may be best turned off for an indoor music program if incorrect classification is found to be an issue.

8.5.6.4 Transient Noise Reduction

This class of processing algorithm is designed to control the loudness of sounds with a very sharp onset, such as cutlery clanging on plates and slamming doors, which can cause substantial output-level overshoot during the amplification algorithm's attack time. Such overshoots can also reduce pleasantness for musical instruments such as the xylophone (Moore et al. 2011). Percussion instruments may potentially trigger these algorithms to apply gain reductions designed to suppress nonmusical sounds and hence create temporal and/or spectral distortions that are deleterious for the perceived timbre of these and other simultaneous instruments. Given the lack of studies on the effect of transient reduction algorithms on music perception, it may be best to err on the side of caution and disable such algorithms for music programs.

8.5.6.5 Reverberation Reduction

Reverberation can cause environmental classification algorithms to classify nonmusic sounds as music (Büchler et al. 2005), although it is unclear whether this could be avoided by processing the classifier input with a reverberation reduction algorithm. It is also unclear whether reverberation reduction algorithms should also process the music signal presented to the aid user. Such algorithms are relatively simple in hearing aids. Typically, they estimate the reverberation time from the input level's maximum rate of decay and use this to estimate the decay of late reflections from previous inputs. The gain is then reduced in channels where the ratio of the direct-to-reverberant sound is low (Roeck and Feilner 2008; Jeub et al. 2012). It is not clear whether the dynamics of music, such as slow decays, can be problematic for some methods of estimating reverberation time. It is also unclear whether reverberation should be reduced when listening to music and whether this depends on the reverberation time, hearing loss, or relative levels of simultaneously played musical instruments. Given the lack of studies, it may be best to err on the side of caution and disable such algorithms to avoid potential negative effects.

8.5.7 *Environmental Classification*

Environmental classification algorithms designed for hearing aids can correctly classify music about 80–90 % of the time, depending on the level of algorithm complexity, and can classify classical music more reliably than pop/rock (Büchler et al. 2005; Gil-Pita et al. 2015). However, noises with a tonal component (e.g., the sound of a vacuum cleaner) and reverberant or amplitude-compressed speech can be misclassified as music, while pop music can be misclassified as noise or speech in noise (Büchler et al. 2005). While classifiers have traditionally been trained by the manufacturer with a large database of stimuli, systems have been investigated that can be trained during everyday use to classify the speech, noise, and music stimuli encountered by the aid user (Lamarche et al. 2010). If an aid user listens only to a narrow range of music genres, such systems could potentially learn to identify the user's favorite music genres more reliably and possibly treat disliked genres as noise that should be suppressed.

However, it is not clear how closely environmental classification algorithms in commercial hearing aids follow the algorithms described in the literature; they may be simplified to meet the constraints of hearing aids resulting in reduced classification accuracy. In this case, incorrect automatic switching of settings could be more problematic than manually changing to a dedicated music program when needed. In addition, music may be a wanted or unwanted sound depending on the listening situation, so environmental classifiers may automatically switch to the music program when music is an unwanted sound. Ultimately, the preference for manual versus automatic switching to and from a music program may be determined by the accuracy of the classifier and the listening habits of the aid user.

8.6 Summary

Hearing aid circuits and signal-processing algorithms have traditionally been designed with speech as the primary signal of interest, which probably explains at least some of the user dissatisfaction with hearing aids when listening to music. The wider range of spectral shapes, levels, and temporal characteristics of music than of speech places greater demands on hearing aid circuits and algorithms. Music is an important signal to most hearing aid users so it is important that hearing aids process music appropriately. However, relatively little research has investigated how well music *is* handled by hearing aids and even less has investigated how music *should be* handled by hearing aids.

It is important that hearing aid circuits can handle the 100-dB dynamic range of music with minimal added noise and distortion. Peak music input levels of at least 105–110 dB SPL need to be passed to the DSP without peak clipping or limiting distortion, and more demanding situations may require higher limits. The required input noise floor during pauses and/or soft music passages has largely been disregarded.

Some solutions to reduce peak clipping result in higher input noise levels and it is not clear whether this interferes with music enjoyment. The required dynamic range of the output and receiver circuits has also received little attention. It is not clear whether output-circuit noise is problematic in quiet listening situations for people with milder hearing losses. Research on preferred bandwidth suggests that typical receivers may not be capable of providing preferred output levels from approximately 5–10 kHz.

If the circuits are capable of passing a clean music signal to the DSP and ear, then the perception of the processed music largely depends on the hearing loss and the DSP algorithms. Aided perception of rhythm, pitch, melody, and timbre is often poorer than normal, although no studies have investigated the relative contributions of hearing aids and hearing loss, and few studies have investigated the effects of different signal-processing algorithms on these percepts. Instead, most research has focused on comparing the overall sound quality or pleasantness of music between different amplification conditions with simulated aids and using fitting prescriptions that incorporated the speech spectrum into their design (i.e., not a special music fitting). In general, these studies have shown that music should be amplified over a frequency range up to about 7.5–10 kHz for people with relatively shallow high-frequency audiogram slopes but perhaps only up to 5–5.5 kHz for people with steeper slopes. A prescription that provides substantially more gain from 4 to 10 kHz (CAM2) is slightly preferable for listening to music to one that provides less high-frequency gain (NAL-NL2). The aided bandwidth should extend down to 90–200 Hz when listening to music, although this is based on receiver roll-off (not the fitted gain).

It is less clear how nonprescribed parameters should be set for music. Although fast WDRC time constants can give quality ratings similar to those of linear gain, paired comparisons reveal that a slower release can be slightly preferred depending on the music genre and input level. Low or high numbers of WDRC channels can be equally preferred in some conditions, although fewer channels can be preferred with faster time constants, higher CRs, poorer frequency selectivity, and/or some music genres. Such interactions between amplification parameters were highlighted by the preference for a 32-channel ADRO aid over a dual 4/15-channel WDRC aid, where the preferred aid had more channels but slower time constants and a different amplification strategy and fitting rationale. It is important that future research investigates interactions among amplification parameters, channel numbers, and types of music as well as the effects of these on music percepts such as rhythm, pitch, melody, and timbre. This will bring us closer to the ultimate goal of fitting methods for music (as opposed to speech) that prescribe parameters such as time constants in addition to frequency- and level-dependent gains.

Studies into the effects of signal-processing algorithms other than amplification on music perception have been scarce, and in general, more are needed to arrive at firm conclusions. Research into the effects of signal path delay with open-canal aids suggests that temporal desynchronization of music above and below 1 kHz by 2–10 ms is not problematic and neither are spectral ripples due to delays of 1–3 ms. However, this is early work and research should be extended to the effects on music percepts with a wider range of stimuli and more realistic, nonlinear, and/or occluding fittings.

Frequency-lowering algorithms are sometimes preferred for music listening and sometimes have significant effects on different musical percepts. It is important that the reasons for this be better understood, so that the benefits of such algorithms can be maximized and the negative effects minimized. In the presence of music, acoustic feedback cancellation algorithms can generate various types of distortions and extraneous sounds and their ability to control acoustic feedback can be compromised. As it is not always practical to turn feedback cancelers off when listening to music (depending on the fitting), there appears to be a general need for improvement. Clinical recommendations have included turning off adaptive and noise reduction features for music where possible. This seems an eminently sensible approach that avoids the distortion of music by algorithms that generally target speech as the desired signal. However, some exceptions that warrant further investigation include using a fixed directional microphone response to improve the music-to-audience ratio, the efficacy of using multiple-channel expansion to suppress input circuit noise but potentially reducing audibility for very soft music, the appropriate use of wind noise reduction when listening to music outdoors, and environmental classifiers that learn to reliably identify their user's favorite music genre(s).

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Chapter 9
Clinical Verification of Hearing Aid Performance

Kevin J. Munro and H. Gustav Mueller

Abstract The general goal of providing amplification is to improve functional auditory capacity and restore good communication skills. Amplification should restore the audibility of soft sounds, provide improved intelligibility of speech at conversational listening levels, and ensure that intense sounds are not amplified to an uncomfortably loud level. There are several prescription methods that provide frequency-specific target values for soft, conversational, and intense sounds. Despite differences in the target values, no validated prescription method has been clearly shown to be superior to any of the other methods in terms of patient benefit (e.g., greater satisfaction, less residual disability). However, clinical studies have clearly shown that when a well-researched prescriptive approach is used and appropriate gain is delivered across frequencies, speech intelligibility is enhanced, and there is improved patient benefit and satisfaction. There is also irrefutable evidence that the audiologist can improve the match to the prescription target values using a probe microphone placed within the patient's ear canal. As a result, carefully conducted verification is an essential component of long-term success with amplification. The most recent generation of prescription methods provides a degree of personalization to the target values beyond that associated with hearing threshold levels. However, there is an urgent clinical need to address the wide range of clinical outcomes that occur in hearing aid users with apparently similar characteristics.

Keywords 2-cc Coupler • Desired sensation level • Long-term average spectrum of speech • Maximum output level • National acoustics laboratory • Probe microphone • Real-ear aided response • Real-ear insertion gain • Real-ear measurement • Real-ear saturation response • Real-ear-to-coupler difference • Speech intelligibility index • Verification

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9.1 Introduction

The hearing aid fitting process can be viewed as a series of sequential steps, as shown in Fig. 9.1. The process commences with an assessment of (1) impairment, (2) hearing ability in different listening situations, (3) the ability to participate in activities that require good hearing, and (4) motivation and expectations. In the clinical setting, impairment is most frequently measured using pure-tone audiometry and speech-in-noise recognition measures. The remaining components of the assessment often involve the use of standardized self-report questionnaires (see Curhan and Curhan, Chap. 2; Akeroyd and Whitmer, Chap. 10). The selection of appropriate amplification will include decisions about the fitting arrangement (e.g., bilateral vs. unilateral), style (e.g., behind-the-ear vs. in-the-ear), coupling (e.g., standard earmold or an "open" fitting), specific features (e.g., the need for a volume control), and amplification characteristics of the device. The detection of environmental sounds, appreciation of music (see Zakis, Chap. 8), and spatial hearing (see Akeroyd and Whitmer, Chap. 7) are also important considerations. The main goal, however, of providing amplification is to improve functional auditory capacity and restore good communication skills. The starting point for achieving this goal is to establish amplification targets for gain/frequency response and output. The hearing aids are then adjusted to match the prescription targets. The match to target is most commonly verified using a probe microphone placed within the patient's ear canal. After the hearing aid has been prescribed and fitted, there may

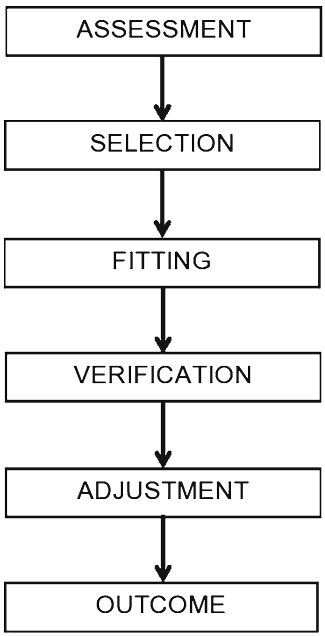

Fig. 9.1 The sequential steps in the hearing aid fitting process [Adapted from Byrne (1981) and Seewald et al. (1996)]

be a need for adjustment and fine tuning to address reports such as lack of clarity, loudness issues, or concerns about the quality of the hearing aid user's own voice.

A variety of clinical tools are available to assess aspects of outcome, including usage, benefit, residual difficulties, and satisfaction (see Akeroyd and Whitmer, Chap. 7). The current chapter is concerned primarily with the use of (1) prescription approaches when fitting hearing aids and (2) probe microphone systems when verifying hearing aid performance in the real ear. For a review of hearing aid prescription approaches and for more detailed information about probe microphone measurements, see Dillon (2012).

9.2 Objectives of the Hearing Aid Fitting Protocol

The hearing aid fitting protocol has three primary objectives: (1) to restore the audibility of soft sounds, (2) to provide improved intelligibility of speech for low and medium input sound levels, and (3) to ensure that intense sounds are not amplified to an uncomfortably loud level. How these objectives are achieved is illustrated in Fig. 9.2. In this figure, the sound pressure level (dB SPL) has been measured in the ear canal close to the tympanic membrane. This is an extremely useful approach because it enables easy visualization of the interrelationship between assessment data, the level of unamplified speech, and the amplification characteristics, which are typically measured in different units and at different reference points. This approach has been encouraged for many years (e.g., DeVos 1968; Erber 1973) and it is one of the key building blocks of the desired sensation level (DSL) fitting method (Seewald 1995), where the chart is referred to as an "SPLogram" (see Sect. 9.4). Similar graphical approaches can now be implemented for other prescription fitting procedures including the National Acoustic Laboratory procedures, where it is referred to as a "speech-o-gram" (Dillon 1999). This approach is available with all probe-microphone equipment for verification purposes, where it often is referred to as "speechmapping."

Figure 9.2a is for a normal-hearing individual. The assessment data include the threshold of hearing (bottom solid line) and the threshold of loudness discomfort, also known as the loudness discomfort level (LDL) or uncomfortable loudness level (top solid line). These represent the lower and upper limits of the individual's dynamic range. Also plotted is the long-term average speech spectrum (LTASS; dashed line). The LTASS is the root-mean-square (RMS) level of a passage of speech, measured in 1/3 octave bands and plotted as a function of band center frequency (Byrne et al. 1994). Such bands approximate the widths of the auditory filters in ears with moderate hearing loss (Moore 2007). The overall level in this example is 65 dB SPL, which is typical of speech at a comfortable conversational level. For simplicity, the 30-dB short-term range of levels of speech in a given band, extending approximately 12 dB above and 18 dB below the RMS level, is not shown. It can be seen that the LTASS lies approximately in the middle of the normal-hearing listener's dynamic range.

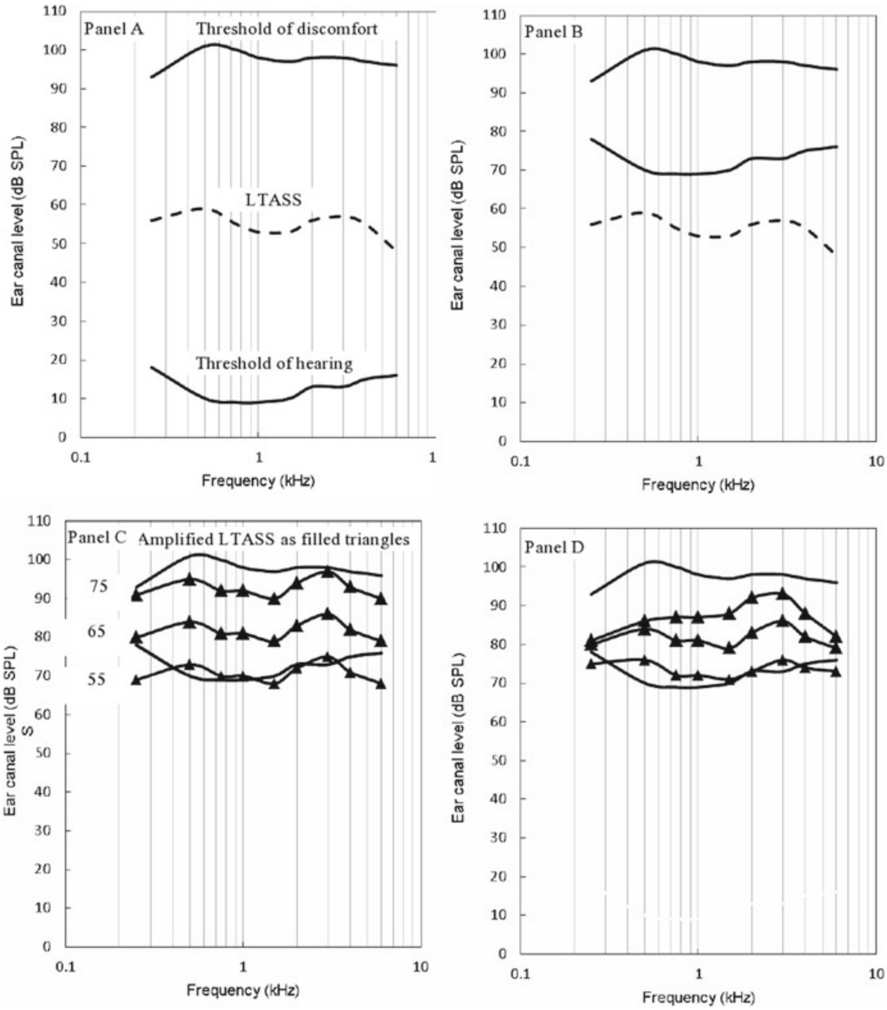

Fig. 9.2 The SPLogram showing the interrelationship between audiometric, acoustic, and electroacoustic variables. The *x*-axis shows frequency (in kHz) and the *y*-axis shows sound pressure level (in dB) in the ear canal. (**a**) SPLogram for a normal-hearing person. (**b**) SPLogram for a hypothetical hearing-impaired patient. (**c**, **d**) The same information as in **b** but includes the effect of appropriately fitted linear hearing aids and multichannel compression hearing aids, respectively, for speech with input levels of 55, 65, and 75 dB SPL

Figure 9.2b is for a hypothetical patient with a flat 60-dB hearing loss. In this example, other than perhaps for a few of the peaks, conversational speech is inaudible. A sensorineural hearing loss increases the threshold of hearing much more than it increases the LDL. Consequently, the dynamic range is much smaller for this patient than for the individual with normal hearing. Not visible on this type of chart is the reduced frequency resolution that accompanies a loss of outer hair cells

(Florentine et al. 1980), although some measures of audibility do include these effects (e.g., Moore and Sek 2013).

Figure 9.2c shows the outcome of providing a hearing aid with linear gain to the hypothetical patient in Fig. 9.2b. The amplified LTASS curves (filled triangles) are shown for input levels of 55 (quiet speech), 65 (normal conversational speech), and 75 dB SPL (loud speech). Amplified conversational speech (65 dB SPL) is now within the dynamic range of the individual. However, in this example, the linear hearing aid does not fully restore the audibility of 55-dB SPL speech at most frequencies, yet it would cause 75-dB SPL speech to be amplified to a level close to the patient's threshold of discomfort (recall that the peaks of speech will have levels up to 12 dB higher than the average line shown in the figure). Figure 9.2d shows the hypothetical outcome of providing an appropriately fitted hearing aid with multichannel amplitude compression. Not only is conversational speech amplified appropriately, but also 55-dB SPL speech is now more audible (because greater gain has been applied for low input levels) and 75-dB SPL speech does not exceed the patient's LDL (because less gain has been applied at high input levels). Placing amplified speech with a wide range of input levels within the patient's residual dynamic range is a goal of all modern prescriptive fitting methods. It can be helpful to quantify the audibility of the unaided and aided LTASS, and this is described in Sect. 9.3.

9.3 Audibility and the Speech Intelligibility Index

The speech intelligibility index (SII; ANSI 1997), formerly known as the articulation index (AI; ANSI 1969), is a method for predicting the intelligibility of speech. The SII is a single value that sums importance-weighted speech audibility across frequencies. It includes a level distortion factor (LDF) to account for the deterioration in performance associated with listening to speech at high sound levels and a desensitization correction to account for the poorer performance of hearing-impaired people relative to normal-hearing listeners. To calculate the SII, frequency-specific information is required about the level of the speech signal and any noise that may be present, along with the hearing thresholds. A value of zero indicates that no speech information is available to the listener and a value of one indicates that all speech information is available to the listener. As the SII value increases, speech understanding generally increases and can be predicted from a transfer function that is specific to the speech material being used during testing. This means that, for a specific SII value, the predicted speech recognition score varies depending on the speech material that is being used. Figure 9.3 shows the relationship between the SII and predicted speech performance for low-predictability material (nonsense syllables, open circles) and high-predictability material (sentences known to the listener, filled circles). For a given SII value, intelligibility is lower for the former than for the latter. For example, an SII of 0.30 would yield about 30% recognition for nonsense syllables and more than 90% for complete sentences.

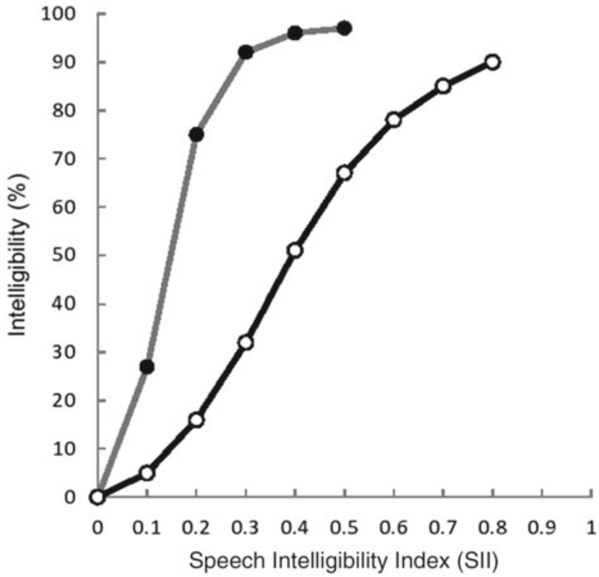

Fig. 9.3 Relationship between the SII and intelligibility for Sentences (*filled symbols* and Nonsense Syllables (*open symbols*) [Adapted from data in Mueller and Killion (1990)]

The accuracy of the SII in predicting speech intelligibility is greater for adults with mild/moderate hearing loss than for those with more severe losses (Magnusson et al. 2001). Also, for the same SII value, speech intelligibility is higher for adults than for children (Scollie et al. 2010a) and for normal-hearing listeners than for hearing-impaired listeners (Ching et al. 1998). The SII also overestimates speech intelligibility for adults or children with cochlear dead regions, which are regions in the cochlea with few or no functioning inner hair cells, synapses, or neurons (Baer et al. 2002; Malicka et al. 2013).

Free software programs are available to assist with calculating the SII (e.g., http://www.sii.to). The SII is also calculated automatically by some clinical probe microphone systems. There are also simple versions of the SII for use in the clinic and these usually involve using a "count-the-dots" audiogram (a graph that shows hearing threshold at standard test frequencies) with hearing thresholds obtained with and without the hearing aid being used. A total of 100 dots are placed within the range of the average speech spectrum at a density representative of the importance function of each specific frequency. Density is greater in the range 1,500–3,000 Hz than at the lower and higher frequency boundaries of the speech spectrum. The SII value is the percentage of the dots that are above the hearing thresholds. Although useful, these simplified versions assume one speech presentation level and no background noise (for an example, see Killion and Mueller 2010).

The SII can be useful when deciding candidacy for hearing aids, when counseling patients, and as an indirect measure of benefit (aided SII minus unaided SII). Bagatto et al. (2011) described one specific application for use in the pediatric clinic. SII values and 95 % range were provided for infants and babies who had been fitted with hearing aids that closely matched the DSL fitting targets (see Sect. 9.4.1). The mean aided SII values decreased from 1.0 to 0.4 as hearing level (HL) increased from 20 dB HL to 90 dB HL, over the frequency range 250–4,000 Hz. For a mean pure-tone average of 60 dB HL the 95 % range of SII values was 0.47–0.8; therefore, children with SII values outside this range can be considered atypical and may require a review of their hearing aid prescription.

The data from Bagatto et al. (2011) also serve as a reminder that the aim is not to ensure an SII value of 1.0 for all patients. A higher SII, which means greater speech audibility, does not necessarily lead to better speech intelligibility because the hearing-impaired ear is less able to extract information from a signal compared to a normal-hearing ear. The high sound levels required to achieve an SII value close to 1.0 may result in hearing loss desensitization, also called reduced effective audibility (Turner and Cummings 1999), and may be associated with excessive loudness. This applies to adults as well as children, particularly those with precipitous high-frequency hearing loss. Attempting to boost the SII by applying high gain at high frequencies may result in a fitting that sounds unacceptable to the patient, who will then reduce the overall gain of the instrument, resulting in insufficient gain over the range 1,000–3,000 Hz; this range makes a strong contribution to intelligibility.

Methods have been developed to quantify the audibility of different types of speech spectra. One example is the Situational Hearing Aid Response Profile (SHARP; Stelmachowicz et al. 1996), which is a computer program that allows calculation of the aided audibility index (AAI) using a variety of different speech spectra. This enables the audiologist to estimate speech audibility for a variety of listening situations, such as when a child is held on the parent/caregiver's hip or in the cradle position, where the level and spectrum of the speech signal may differ from those used during clinical verification of the amplification characteristics. These data can be useful when counseling the family and discussing the impact of different listening situations on the child. SHARP has recently been updated to reflect recent advances in hearing aid signal processing, such as multichannel amplitude compression and frequency lowering (Brennan and McCreery 2014). Copies of SHARP can be obtained by e-mailing audiosharp@boystown.org.

9.4 Prescribing Hearing Aid Gain and Output

Because the degree and configuration of hearing loss vary from one person to the next, it is not appropriate to provide the same amplification to all hearing-impaired patients. It is also not appropriate to provide an arbitrary amount of amplification. The prescribed amplification should be based on individual information obtained from the patient. Amplification characteristics are most commonly selected using pure-tone hearing thresholds, probably because these are standardized and relatively quick and simple to measure. In some hearing aid prescription methods, suprathreshold measures such as loudness judgments are also recommended or required.

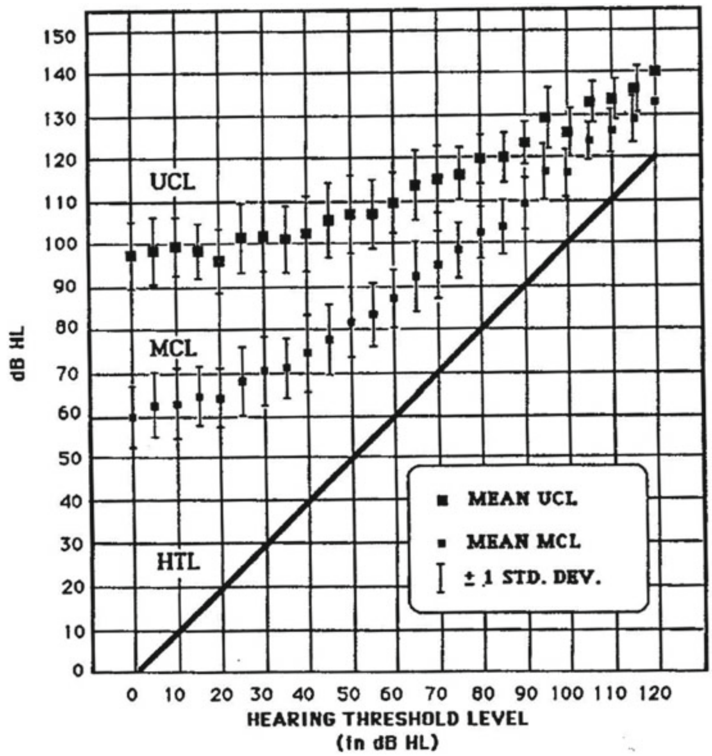

Fig. 9.4 Relationship between hearing threshold, most comfortable loudness level, and uncomfortable loudness level (UCL) [From Pascoe (1988). Reprinted with permission]

The literature on prescription approaches is as confusing as it is voluminous. There are many different prescription methods, with significant variations in target values for the same hearing loss, and little evidence that one method results in greater benefit to the patient than another.

One of the first prescription approaches involved providing frequency-specific gain that equaled the hearing loss (Knudsen and Jones 1935). This "mirroring of the audiogram" means that a patient with, for example, a hearing threshold level of 60 dB HL at 1,000 Hz, would be prescribed 60 dB of gain at this frequency. We know today that for a patient with a sensorineural impairment, this amount of gain would be excessive due to his or her reduced dynamic range and abnormal loudness growth. This is illustrated in Fig. 9.4 (from Pascoe 1988), which shows the relationship between the threshold of hearing, the most comfortable listening level (MCL), and uncomfortable loudness level (UCL). When the hearing threshold level is 0 dB, the UCL is close to 100 dB HL and the MCL is 60 dB above the hearing threshold level, corresponding to 60 dB sensation level (SL). However, the dynamic range reduces with increasing hearing loss such that when the hearing threshold level is 100 dB HL, the UCL is 125 dB HL and the MCL is about 17 dB SL.

Lybarger (1944) reported that, for linear hearing aids, the amount of gain selected by adult patients with sensorineural hearing loss was approximately half of the hearing threshold level. This “half-gain rule” would suggest that our hypothetical patient with a hearing threshold level of 60 dB HL at 1,000 Hz should be prescribed 30 dB of gain. Many of the later prescription approaches for linear aids are derived from this half-gain rule, including Berger’s method (Berger et al. 1979); prescription of gain and output (POGO; McCandless and Lyregaard 1983); Libby (1986); and the National Acoustics Laboratories method (NAL; Byrne and Tonnison 1976; Byrne and Dillon 1986; Byrne et al. 1990).

The formulas for a variety of generic prescription approaches have been incorporated into hearing aid manufacturers’ fitting software and clinical real-ear measurement systems. Most major manufacturers also have proprietary prescriptive methods as part of their software. The term “proprietary” refers to a fitting method unique to a given manufacturer, although it may be a modification of a validated method. These methods often change when different products are released and, unfortunately, they cannot be individually verified, as no real-ear fitting targets exist. Manufacturers’ proprietary prescriptive approaches typically have not been validated or subjected to the same scientific scrutiny as generic approaches. Studies have compared generic and proprietary prescription methods and revealed significant differences in amplification target values for adults (e.g., Keidser et al. 2003; Leavitt and Flexer 2012) and children (e.g., Seewald et al. 2008). In general, these methods are geared toward “first acceptance” rather than audibility and intelligibility. Current hearing aid fitting guidelines (e.g., American Academy of Audiology; American Speech-Language-Hearing Association; British Society of Audiology; International Collegium of Rehabilitative Audiology; and International Society of Audiology) recommend the use of generic validated prescription methods.

9.4.1 Linear Amplification

All linear gain-based fitting procedures provide a single-target frequency-dependent gain curve for a specific audiometric hearing loss configuration. Although linear hearing aids have largely been superseded by multichannel compression devices and multichannel compression prescription methods, the principles behind the development of linear prescriptions also apply to multichannel compression prescription methods. Therefore, two of the most widely accepted methods of prescribing gain for linear hearing aids are summarized in Sects. 9.4.1.1 and 9.4.1.2.

9.4.1.1 NAL Linear Prescription Targets

The prescriptive approaches developed at NAL have been widely adopted in the fitting of hearing aids to adults. The aim of the NAL procedure is to maximize speech intelligibility at the listening level preferred by the patient.

Table 9.1 Use of the NAL-R prescription approach to obtain insertion gain target values for a hypothetical patient with a high-frequency hearing loss

1	Audiometric frequency (kHz)	0.25	0.5	1	2	3	4	6
2	Hearing threshold level (dB) of hypothetical patient	30	35	35	50	50	60	80
3	Constant of 0.15 × AVE HL (0.5, 1, and 2 kHz)	6	6	6	6	6	6	6
4	0.31 × hearing threshold level	9	11	11	16	16	19	25
5	Constant to make speech equally loud	–17	–8	1	–1	–2	–2	–2
6	NAL-R target insertion gain (dB) (3 + 4 + 5)	–2	9	18	21	20	23	29
7	Convert insertion gain (line 6) to 2-cc coupler target	13	12	12	16	21	21	11

The hearing threshold levels of the patient are shown in the second row. The insertion gain targets in row 6 are obtained by summing rows 3, 4, and 5 AVE.

The original NAL formula was developed by Byrne and Tonnison (1976). It was based on the observation that the preferred gain (aided minus unaided response in the ear canal) at 1,000 Hz, in experienced adult hearing aid users, was 0.46 of the hearing threshold level at 1,000 Hz, which is close to the half-gain rule. The assumption was made that the gain at every frequency should be 0.46 of the hearing threshold level. Two sets of corrections were then applied to account for (1) the shape of the LTASS (e.g., less gain is needed at low frequencies where the speech level is highest) and (2) differences in loudness across frequency for a fixed overall level. Dillon (2012) makes the important point that the basis of the formula (although no longer used) is still relevant today.

Evaluation of the original NAL formula during the 1980s revealed that it did not quite achieve equal loudness at all frequencies. The formula was revised but retained the “almost” half-gain rule for the three-frequency average (500, 1,000, and 2,000 Hz). This revised fitting method is known as NAL-R (Byrne and Dillon 1986).

A worked example of how gain is calculated using NAL-R is shown in Table 9.1. The hearing threshold levels of a hypothetical patient with a high-frequency hearing loss are shown in row 2. The prescription targets in row 6 are obtained by summing rows 3, 4, and 5. Row 3 is a constant that is based on 0.15 of the three-frequency average (500, 1,000, and 2,000 Hz). Row 4 is 0.31 of the hearing threshold level at that frequency. Row 5 is a constant that is independent of the hearing threshold level but adjusts the gain to make speech equally loud in each frequency band.

Because the initial selection of an appropriate hearing aid is based on data from the manufacturer that is obtained using a 2-cc coupler, and not a real ear, the target value in a 2-cc coupler requires another correction factor (row 7). For example, at 4,000 Hz, the correction factor when determining the desired 2-cc coupler target is +21 dB for a behind-the-ear hearing aid; therefore, the target value for full-on gain in a 2-cc coupler is 44 dB at 4,000 Hz. This 21 dB value includes 15 dB of reserve gain, which is applied to all frequencies. The remaining 6 dB is to correct between the real-ear insertion gain and 2-cc coupler gain due to microphone location effects and differences between the acoustic properties of the occluded ear and the coupler and is known as the coupler response for flat insertion gain (CORFIG; Killion and Monser 1980).

Evaluations of NAL-R during the early 1990s revealed that patients with a profound hearing loss preferred more gain at low frequencies (where they typically had more "usable" hearing) and less gain at high frequencies. In addition, when the three-frequency average was above 60 dB HL, it was found that more gain was needed and the 0.46 gain rule was increased to 0.66. This resulted in a further revision of the prescription procedure and became known as the NAL-RP prescription (Byrne et al. 1990).

So far, the discussion of the NAL prescription methods has been limited to prescribing the gain-frequency response. While a patient may use a hearing aid even if the gain-frequency response is inappropriate, there is a danger that they will not use the hearing aid at all if the maximum output of the aid exceeds their LDL. The "ideal" maximum output level will prevent excessively high output levels without introducing distortion and reducing speech intelligibility. Some prescription approaches avoid excessively high output levels by setting the maximum output a little below the patient's LDL (measured directly or predicted from the hearing thresholds). Other procedures, including NAL, also consider the minimum acceptable limit, assumed to be that which causes only a small amount of limiting with a speech input level of 75 dB SPL. The approach taken by NAL is to provide maximum output targets that are midway between the LDL and the highest output level that avoids excessive saturation for a 75-dB SPL speech input level (Dillon and Storey 1998). This approach to setting maximum output has been validated in studies by Storey et al. (1998) and Preminger et al. (2001).

9.4.1.2 Desired Sensation Level Linear Prescription Targets

The aim of the DSL prescriptive approach, developed by Seewald and colleagues from the University of Western Ontario, is to define gain-frequency response characteristics that make amplified speech audible, comfortable, and physically undistorted in each frequency region (Seewald et al. 1985, 1987; Ross and Seewald 1988; Seewald and Ross 1988).

The DSL approach has been widely adopted to fit hearing aids to infants and children because it is the only prescriptive approach that, by design, specifically accounts for factors that are associated with provision of amplification for this population. For example, the LTASS used in the DSL approach was derived using a microphone placed at the entrance of the ear canal of the child, in addition to a reference position in the sound field, because self-monitoring is required for speech and language development (Cornelisse et al. 1991). Additionally, it is appropriate to provide greater aided audibility for a younger child who is developing speech and language than for an older child or an adult who has already developed speech and language.

Early versions of DSL used tables of values that specified target sensation levels for amplified speech as a function of frequency and hearing level. These DSLs were based on data describing the speech sensation levels that were associated with comfortable listening levels for different amounts of hearing loss (e.g., Pascoe 1978)

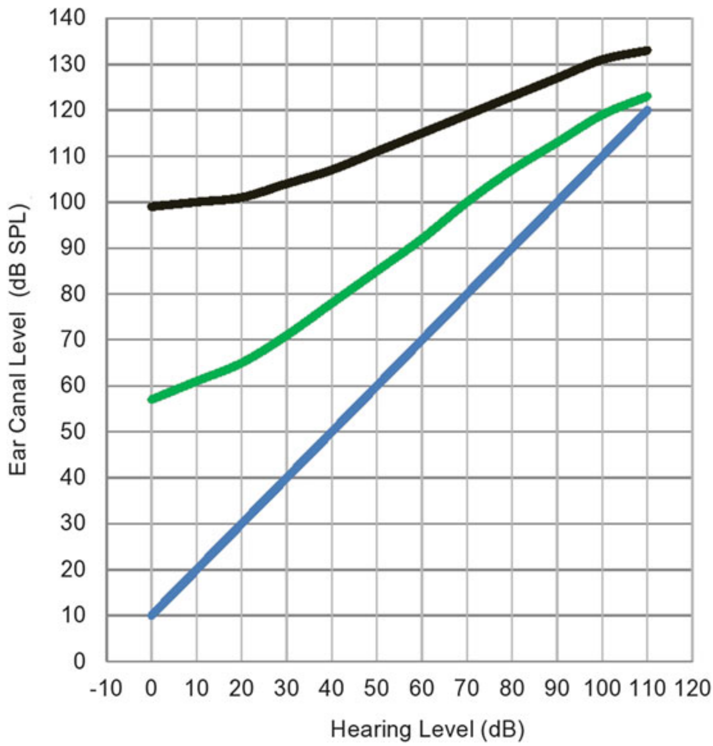

Fig. 9.5 Desired ear canal target levels, at 1,000 Hz, for amplified speech (*green line*) and maximum output (*black line*) (dB SPL) as a function of hearing threshold level (Based on tabulated data from DSL 3.0 user's manual)

and levels that led to ceiling speech recognition performance for children with sensorineural hearing impairment (e.g., Gengel et al. 1971). The target 1/3 octave output levels for aided speech are placed one standard deviation below the mean MCL for pure tones, and the target output levels for maximum output are placed one standard deviation below the mean LDL values (to reduce the possibility of the infant having a hearing aid whose output exceeds the LDL). The relationship between hearing threshold level, amplified speech level, and maximum output for a center frequency of 1,000 Hz is illustrated in Fig. 9.5. This shows that, for example, if the hearing threshold at 1,000 Hz is 35 dB HL, the target level for amplified speech is 74 dB SPL (or 30 dB SL) and the target level for maximum output is 105 dB SPL. This procedure is applied at each center frequency. The required gain is then calculated by subtracting the speech input level from the target output level.

The linear version of the DSL prescriptive approach was first implemented as a software version in 1991 (v3.0; Seewald et al. 1991). In addition to recommending target values, it included features associated with differences in the acoustic properties of the external ear and used the SPLogram format to assist with visualization of the precise relationship between the dynamic range of the individual, unamplified speech levels, and the recommended amplification characteristics.

9.4.2 Multichannel Compression Prescription Targets

The spectral characteristics and overall level of speech can vary depending on a number of factors, including vocal effort of the speaker, distance from the listener, and signal-to-noise ratio (SNR; see, e.g., Pearsons et al. 1977). In addition, the reduced dynamic range and abnormal loudness growth associated with a sensorineural hearing loss mean that the desired gain-frequency response changes with input level. Prescription methods for hearing aids with amplitude compression provide gain targets for several input levels.

Most multichannel compression prescription approaches are based on the goal of providing overall loudness that is similar to that experienced by a normal-hearing individual. For some prescription methods (e.g., NAL-NL1 and NAL-NL2), sounds that are described as “soft” by a normal-hearing listener should be perceived as soft by the hearing-impaired patient listening via hearing aids and sounds that are described as “loud” by a normal-hearing listener should also be perceived as loud by the hearing-impaired patient. Other prescription methods (e.g., DSLm[i/o] and CAMEQ2-HF) take a different approach that can lead to soft sounds having greater-than-normal loudness because priority is given to restoring speech audibility. There is generally good agreement between linear and multichannel compression target values for speech at conversational levels; however, with multichannel compression, more gain is applied for low input levels and less for high input levels.

9.4.2.1 NAL-NL1 and NAL-NL2 Multichannel Compression Prescription Targets

The multichannel compression versions of the NAL prescription method continue to be threshold-based procedures. Whereas the linear versions of the NAL method attempted to restore normal loudness to the hearing-impaired patient for medium input levels, the multichannel compression methods aim to maximize speech intelligibility for a given overall loudness. Like all multichannel compression fitting methods, the NAL methods provide targets for multiple input levels (e.g., 50, 65, and 80 dB SPL). For gently sloping high-frequency hearing loss, the gain targets for a 65 dB SPL input level are similar for the linear and multichannel compression methods.

The first NAL multichannel compression method, NAL-NL1, appeared at the turn of the century (Dillon et al. 1999; Byrne et al. 2001). The derivation of NAL-NL1 targets involved two theoretical models: one based on the SII, in which an allowance was made for the effects of hearing loss desensitization, and the other based on calculation of loudness (Dillon 1999). Many audiometric configurations and speech input levels were used to determine the gains required to give maximum speech intelligibility without exceeding normal loudness. The formula used to derive the prescription targets is not in the public domain.

Clinical evaluation of speech intelligibility for normal-hearing and hearing-impaired adults revealed that the greater the hearing loss, the greater the tendency of the SII to overestimate speech intelligibility. The NAL-NL1 derivation process was repeated using revised SII and loudness models. A variety of psychoacoustic parameters, including information about frequency regions with no auditory function, called cochlear dead regions (Moore 2004), were also used when calculating effective audibility; however, these did not explain much of the discrepancy between observed and predicted speech intelligibility. This resulted in a revised multichannel compression prescription method called NAL-NL2 (Keidser and Dillon 2006; Keidser et al. 2012). Relative to NAL-NL1, the revised formula prescribes relatively more gain at low and high frequencies and less gain at midfrequencies. NAL-NL2 also attempts to personalize the prescription so that two individuals with the same pure-tone thresholds may have different recommended target settings (see Sect. 9.4.3). More detailed information about the NAL-NL2 fitting method can be found at www.nal.gov.au.

9.4.2.2 DSL[i/o] AND DSLm[i/o] Multichannel Compression Prescription Targets

With the advent of multichannel compression hearing aids, DSL was revised and updated to DSL[i/o] and implemented in software version 4.0 (Cornelisse et al. 1995). This prescribes targets for amplified speech and output limiting for use with both linear and multichannel compression hearing aids (and compression ratios for the latter). The target sensation levels for conversational speech are equivalent to those recommended by earlier versions of the DSL prescription approach. DSL[i/o] actually has two procedures. The procedure that has been predominantly used by audiologists is called DSL[i/o] curvilinear. The alternative procedure was known as DSL[i/o] linear but "straight compression" may be a better term because the compression ratio is constant over most of the dynamic range.

DSL[i/o] underwent revision and was replaced with DSLm[i/o], often referred to as DSLv5.0 (Scollie et al. 2005). One modification is that compression is applied over a smaller range of input levels. The algorithm includes four stages (the "m" in the name stands for multistage) to reflect the signal processing in modern digital hearing aids (see Launer, Zakis, and Moore, Chap. 4). In order of increasing level, the stages are (1) amplitude expansion, (2) linear gain, (3) amplitude compression, and (4) output limiting. Relative to DSL[i/o], DSLm[i/o] prescribes similar gain for a flat hearing loss and slightly less gain at low frequencies for more severe or steeply sloping hearing losses.

DSLm[i/o] provides target values based on electrophysiological measures of hearing threshold levels for babies when it has not been possible to obtain a reliable behavioral measure of hearing thresholds. DSLm[i/o] has revised normative data for differences in the acoustic properties of the external ear for different age groups, provides target values with less gain when listening in a noisy environments, corrects for conductive hearing loss (which is associated with a greater dynamic range than sensorineural hearing loss), and prescribes an optional 3-dB reduction in gain to allow for binaural loudness summation when fitting bilateral hearing aids to adults. More detailed information about the DSLm[i/o] fitting method can be found at www.dslio.com.

9.4.3 Personalization of Prescription Targets

In the field of healthcare, there is increasing emphasis on personalization of treatment. This is done with the intention of accounting for individual variations to improve treatment outcomes. Hearing aid prescriptions are based on the audiometric profile of the individual so they already contain an element of personalization. However, recent multichannel compression fitting methods take this personalization a step further by accounting for additional factors such as listening experience, gender, language, and listening environment.

One example of personalization is the difference in prescription between typical children (longstanding congenital hearing loss) and typical adults (hearing loss acquired as they pass along the lifespan). The NAL-NL1 prescription was found to provide about 3 dB too much gain for adults with mild-to-moderate hearing loss listening at average input levels (Keidsler et al. 2008). On the other hand, children were found to prefer around 2 dB more gain than prescribed by NAL-NL1 (Ching et al. 2010a). To accommodate this, the NAL-NL2 method prescribes about 5 dB less gain for adults than for children with the same audiogram.

A key feature of DSLm[i/o], published a few years before NAl-NL2, is the inclusion of separate target values for children (with a congenital hearing loss) and adults (with an acquired hearing loss). The reason for separate targets is that adult hearing aid users, especially new users, prefer aided listening levels below those for hearing-impaired children (Scollie et al. 2005). Thus, the prescription formula is based on "preference" in addition to audibility and loudness. The mean difference between target gains for children with a congenital hearing loss and adults with an acquired hearing loss is about 10 dB for mild-to-moderate hearing losses and 3 dB for more severe losses. The DSLm[i/o] method also provides different target values when listening in a quiet environment and listening in noise, and target values change if the individual's measured LDLs are entered. That is, given that the LDL determines the upper limit of the dynamic range, the targets for soft, average, and loud sounds shift higher or lower to account for the newly calculated dynamic range for that individual.

Because males have been shown to prefer more gain than females (Keidser and Dillon 2006), NAL-NL2 prescribes 2 dB more gain for males. Experienced hearing aid users have also been shown to prefer more gain than new hearing aid users (Keidser et al. 2008). The difference in NAL-NL2 target gains for the two groups increases from 0 dB for mild hearing losses up to 10 dB for more severe losses.

9.4.4 Comparing Prescription Target Values

Prescription methods in addition to those described above include the Cambridge loudness equalization method (CAMEQ; this aims to give equal loudness for all frequency bands in the range 0.5–4 kHz; Moore et al. 1999a, b) and the Cambridge restoration of loudness method (CAMREST; this aims to restore both loudness and timbre to “normal”; Moore 2000). Given the plethora of prescription methods, there are two important questions: (1) Do they result in different amplification targets, and (2) if so, does it matter? There have been several comparisons of prescription methods and these show that, depending on audiometric configuration and speech input level, prescription targets can vary in overall gain and in the shape of the frequency response curve. One example, from Seewald et al. (2005), is shown in Fig. 9.6. In this example, Seewald compared the prescription target values for three multichannel compression fitting methods: CAMFITv1 (CAMREST), NAL-NL1.2.8, and DSL v4.1a. The prescription targets are displayed as real-ear aided gain (REAG) for pure-tone input levels of 50, 65, and 80 dB SPL. The pure-tone gains for the various Cambridge methods depend on the number of compression channels. The figure indicates 2-channel for NAL-NL1, but it does not indicate the number of channels for CAMFIT. The gains prescribed for speech by the Cambridge methods do not depend on the number of channels.

The top panel shows the targets for an audiogram that slopes gently downward from 30 dB HL at 0.25 kHz to 50 dB HL at 4 kHz. The middle panel shows the targets for an audiogram that slopes gently downward from 50 dB HL at 0.25 kHz to 70 dB HL at 4 kHz. The bottom panel shows the targets for an audiogram that slopes gently downward from 70 dB HL at 0.25 kHz to 90 dB HL at 4 kHz. For these examples, the gain/frequency response target values are similar in shape; however, there are significant differences in gain among the three methods, especially for the lowest (50 dB SPL) input level. For hearing aids with a volume control, the adult patient can adjust the gain. This, however, is not possible for infants and children who lack the developmental capacity to use a volume control. A single volume control, of course, cannot address differences in the shape of the frequency response among the three methods.

A recent comparison of modern prescriptive methods was reported by Johnson and Dillon (2011). These authors compared the NAL-NL1, NAL-NL2, DSLm[i/o], and CAMEQ2-HF (Moore et al. 2010) for several audiometric configurations. They calculated prescribed gain, overall loudness, effective audibility, and predicted speech intelligibility for an average conversational input level. NAL-NL2 and DSLm[i/o] provided comparable overall loudness for a 65-dB SPL ILTASS input, and these methods also provided comparable predicted speech intelligibility in quiet

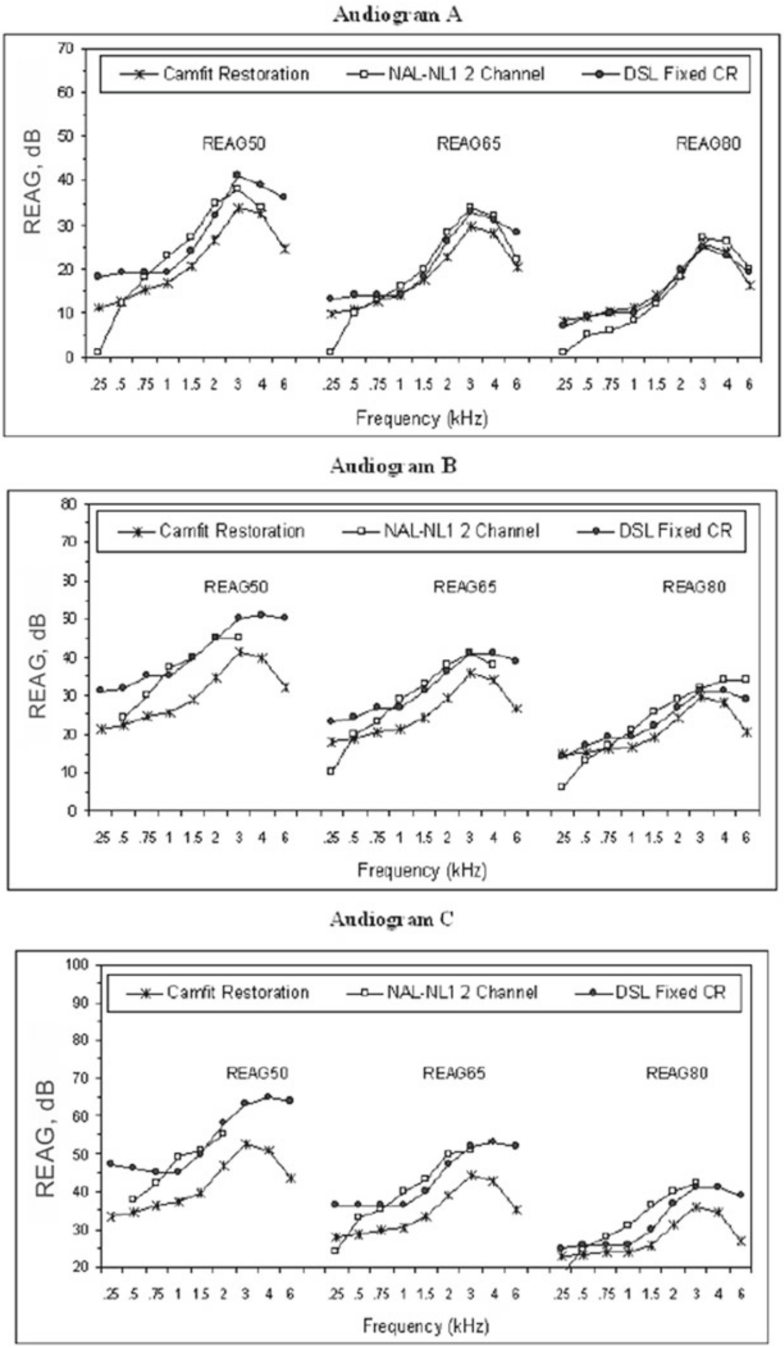

Fig. 9.6 A comparison of real-ear aided gain, as a function of frequency, for three audiometric configurations using three multichannel compression prescription methods: CAMFIT restoration (CAMREST), DSL 4.1 fixed CR, and NAL-NL1. See text for details [From Seewald et al. (2005). Reprinted with permission.]

and noise. CAMEQ2-HF provided a greater average loudness, similar to that for NAL-NL1, with greater audibility of high frequencies but no significant improvement in predicted speech intelligibility. The authors concluded that prescribed insertion gains differed across the prescriptive methods, but when these differences were averaged across the different hearing losses, they were negligible with regard to predicted speech intelligibility at normal conversational speech levels.

When fitting hearing aids to adults, most clinicians are faced with the decision of choosing between NAL-NL2 and DSLm[i/o], as these are the most common methods in manufacturers' fitting software and also the verification targets that are available on most probe-microphone equipment. Johnson (2012) directly compared these two methods, illustrating that for many patients, the fittings will be very similar, although differences of 5 dB or more can occur depending on the degree of hearing loss and audiometric configuration.

Although research has shown real-world advantages of using a validated prescriptive method compared to proprietary fittings (Abrams et al. 2012), there have been relatively few well-designed studies that have compared outcomes with the different validated prescription methods. In addition, few of the studies used blinding and this raises concerns about possible biases (see, e.g., the studies on placebo and expectations in hearing aid fittings by Dawes et al. 2011, 2013). Early studies have generally shown that hearing aid users who are experienced with a particular prescription tend to prefer the familiar prescription to alternative prescriptions (e.g., Ching et al. 1997; Scollie et al. 2000). It is possible that this reflects auditory acclimatization to the familiar prescription. Moore et al. (2001) compared the effectiveness of CAMEQ, CAMREST, and DSL[i/o] for 10 experienced adult hearing aid users. Performance on self-report and laboratory measurements of speech recognition for sentences presented in quiet and in steady and fluctuating background noise was measured after the gain was adjusted from the prescription target (to achieve user acceptability). Mean performance did not differ significantly for the three procedures, although the gain adjustment reduced the differences between the three prescription methods. Although smaller gain adjustments were required for the CAMEQ and CAMREST methods than for DSL[i/o], this may have been because these prescriptions differed less from the patients' typical gain settings. However, in a study using a similar methodology to Moore et al. (2001), Marriage et al. (2004) compared the same three procedures in a group of new and a group of experienced hearing aid users and obtained similar results.

In a collaboration between the developers of the NAL-NL1 and DSLv4.1 prescriptions, performance was compared for a group of 48 school-age children with up to moderately severe hearing losses, using a double-blind crossover design. The study evaluated both efficacy (differences in a laboratory setting) and effectiveness (real-world outcomes). The findings of this comprehensive and well-designed study (Ching et al. 2010a, b, c; Scollie et al. 2010b, c) revealed that hearing aid benefit was high when either of the two prescriptions was used. Interestingly, the findings suggested that to achieve optimal audibility of soft speech, children need more gain than was prescribed by NAL-NL1, but to achieve listening comfort in noisy places, they needed less gain than was prescribed by DSLv.4.1 (Ching et al. 2010a). There were some differences between methods regarding overall preference following real-world experience, which were country specific (Canada vs. Australia). These findings are consistent with a bias in preference toward the prescription to which the children were accustomed.

Moore and Sek (2013) compared preference judgments for NAL-NL2 and CAM2 (initially called CAMEQ2-HF, a modified and extended version of CAMEQ with gain targets for frequencies up to 10,000 Hz; Moore et al. 2010). Participants were 15 adults with mild, sloping sensorineural hearing loss, typical of candidates for wide-bandwidth hearing aids. Judgments of overall sound quality were obtained for male and female speakers and for different types of music. Judgments of speech clarity were obtained for male and female speakers listening in a background of speech-shaped noise and in a background of male or female speakers. CAM2 was preferred for overall sound quality and the clarity of speech in noise. Further work is needed to determine if these laboratory-based findings also apply to real-life listening environments. Often, in real-world studies, the limitations of the hearing aids' bandwidth and the need to avoid acoustic feedback mean that differences in gain-frequency response between prescription methods become minimal.

9.5 Real-Ear Probe Measurements

Once a hearing aid has been selected for the patient, the audiometric information entered into the hearing aid fitting software and the prescriptive method selected; the hearing aid manufacturer's fitting software generates a prescription target and programs the hearing aid accordingly. However, the match to target requires verification in the ear canal of the individual hearing aid user because the initial match is based on average correction factors (including the effects of the hearing aid microphone location and the acoustics of the ear canal and earmold). Moreover, it is possible that the manufacturer has altered the fitting to deviate from the original prescriptive version. For example, Aazh et al. (2012a) reported that for 51 different hearing fittings, all programmed using the manufacturer's NAL-NL1 software, only 29 % of the fittings were within 10 dB of prescriptive targets over the range 250–4,000 Hz. This failure rate and the degree of error (>10 dB) are much higher than would be expected simply because of transfer functions that deviated from average. This finding has been extended to frequency response slope and to a range of input levels (Munro et al. 2016)

Probe microphones allow the sound level to be measured in the ear canal, which is the only way to verify a prescriptive fitting method. Clinically friendly real-ear measurement systems were first introduced into clinical practice in the early 1980s (Mason and Popelka 1986) and are now standard equipment in many audiology clinics. Real-ear measurement terminology and procedures are covered by national and international standards (e.g., ANSI S3.46-2013; IEC 61669 2001).

9.5.1 Equipment and Procedural Issues

A real-ear measurement system comprises a microphone coupled acoustically to the ear canal by a fine-bore flexible tube. The nonflat frequency response of the probe tube is allowed for by calibrating it against a reference microphone that has a flat response. This calibration does not require the hearing aid user to be present. The correction is stored in the real-ear measurement system and applied to any subsequent real-ear measurements. It must be repeated when the tube is changed. Some manufacturers (e.g., Frye Electronics) use a somewhat different procedure that does not require tube calibration.

Most real-ear measurement systems have two microphones: the probe tube microphone that measures the sound level in the ear canal and a control/reference microphone that is placed close to the hearing aid user's ear and measures the sound level generated by the loudspeaker. Movement of the patient, relative to the loudspeaker, will alter the sound level reaching the ear and the hearing aid microphone. To reduce this measurement error, the output from the control microphone is used to compensate for any change in level by increasing or decreasing the electrical signal delivered to the loudspeaker (Fig. 9.7) during the measurement procedure.

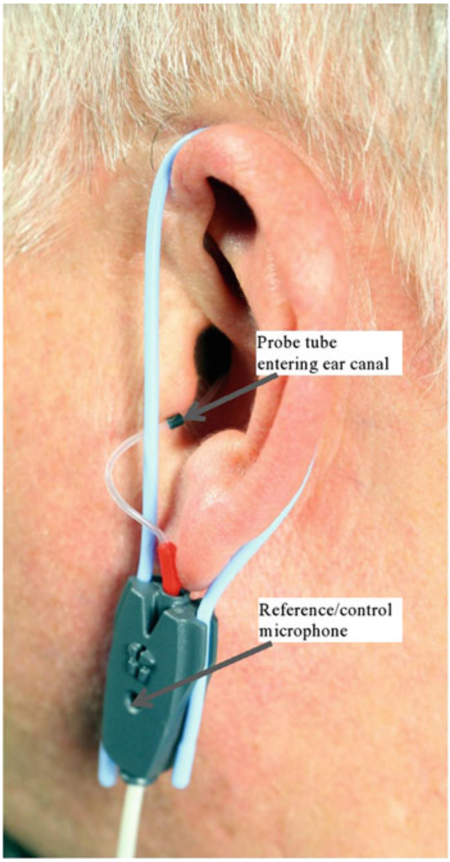

Fig. 9.7 The probe tube is in the ear canal and the reference/control microphone is below the pinna with the outlet facing outwards [From Verifit® User's guide v 3.10. Reprinted with permission.]

This is referred to (rather too clumsily for everyday use) as the “modified pressure method with concurrent equalization.” Ideally, the control microphone should be located as close to the hearing aid microphone as possible (it typically hangs near the ear lobe) to maintain a constant sound level at the hearing aid microphone. Substantial differences in sound level can occur depending on the location of the control microphone on the head (Feigin et al. 1990; Ickes et al. 1991; Hawkins and Mueller 1992).

Sometimes it is necessary to undertake real-ear measurements without the benefit of a control microphone. One example is when the hearing aid is being used with an “open fitting” (i.e., a nonoccluding earmold). In this situation, amplified sound may leak out of the ear canal and back to the control microphone. The sound level from the ear canal may be higher than the sound produced by the loudspeaker when measured at the control microphone, and this causes the system to reduce the actual loudspeaker output. This leads to an underestimate of gain (the artificially recorded high input level at the control microphone minus the output level measured by the probe microphone in the ear canal; see review by Mueller and Ricketts 2006). In this case, it may be necessary to disable the control microphone (cf. Aazh et al. 2012a). If the control microphone is disabled, a calibration process called the “modified pressure method with stored equalization” is performed before the real-ear measurement is made. However, with a disabled control microphone, it is absolutely essential that the hearing aid user (and the audiologist) does not move during the measurement procedure (about 5–8 s), as any resulting change in sound level reaching the hearing aid microphone will not be corrected.

Positioning the patient for testing is important. It is common to use a loudspeaker at 0° azimuth and elevation, with the patient seated about 1 m from the loudspeaker. It is important to keep the patient close to the sound source as this improves the SNR at the ear, which is critical for testing at low levels. It also helps to prevent overdriving the loudspeaker when high levels are presented. There is some evidence to suggest that test–retest reliability is slightly better for a 45° than for a 0° azimuth (Killion and Revit 1987), although this has not always been replicated (Stone and Moore 2004) and, in any case, this minor advantage is outweighed by the practical convenience of the 0° location.

Probably the most important procedural aspect of ear canal measurement is the insertion depth of the probe tube. The probe tube should be positioned as close to the tympanic membrane as possible to obtain valid measures, especially at high frequencies. Figure 9.8 shows the difference in sound level between the tympanic membrane and the probe tip as a function of distance of the probe tip from the tympanic membrane. Because of standing waves, the difference is greatest for high frequencies, with their relatively short wavelength. For example, when the probe tip is 10 mm from the tympanic membrane, the difference is about 6 dB for an 8-kHz stimulus. As long as the tip of the probe tube is within 5 mm of the tympanic membrane, the measured level is accurate to within 2 dB for frequencies below 6 kHz.

There are several procedures for determining an appropriate probe tube insertion depth (Dirks et al. 1996; Mueller 2001) and information about each approach is provided in Annex B of ANSI S3.46-2013 (2003). The most common approach is to

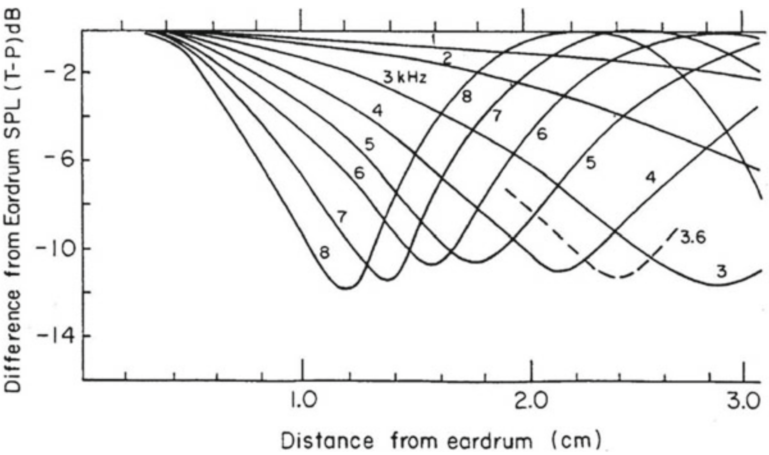

Fig. 9.8 Difference in sound level between the tympanic membrane and the probe tube tip as a function of distance from the tympanic membrane [From Dirks and Kincaid (1987). Reprinted with permission.]

use a fixed insertion depth based on knowledge about typical ear canal dimensions. The length of the average adult ear canal is 25 mm (Zemplenyi et al. 1985) and the typical distance from the ear canal entrance to the external ear (usually measured from the intertragal notch) is 10 mm. Therefore, an insertion depth of 30 mm will place the probe tip within 5 mm of the tympanic membrane for the average adult. If the primary interest is not the absolute level in the ear canal but the difference between the levels for the aided and unaided conditions, the exact location of the probe is not as critical (it must extend 4–5 mm beyond the tip of the hearing aid or earmold), as long as the probe tip is in exactly the same location for the two measurements.

It is important to use a stimulus that will make the hearing aid operate in a realistic manner. Over the years, test stimuli for real-ear measurement systems have ranged from sweep-frequency pure tones to constant-level noise and, more recently, stimuli with similar frequency, intensity, and temporal characteristics to real speech. Information about the characteristics of acoustic stimuli used for real-ear measurements is provided in Annex A of ANSI S3.46-2013 (2013). Various hearing aid features, including digital noise reduction and acoustic feedback suppression, can influence the measured gain when artificial signals are used. An alternative is to use real speech as the test signal. All of today's clinical systems provide recorded and calibrated real speech as a test stimulus. This has the major advantage that multichannel compression operates in a manner that is more realistic than when non–speech-like stimuli are used. The test speech signals differ across real-ear systems. For example, Frye Electronics has a "DigiSpeech" signal and the Audioscan Verifit has a calibrated male talker signal with the LTASS. However, one signal that can be found on all real-ear systems is the international speech test signal (ISTS). This signal is based on speech segments from female talkers of different languages. It has speechlike acoustic properties but is not intelligible (see Holube et al. 2010).

Because the speech material differs across real-ear systems, the ear canal target levels, displayed by the manufacturer's software, need to be adjusted accordingly. Ricketts and Mueller (2009) compared deviations from NAL-NL1 prescription targets for three different popular probe microphone systems using the same hearing aids. Although the deviations were similar across the three systems, generally within 5 dB, they were not identical. Real and potentially clinically significant differences exist across systems. It is currently unclear if the differences were due to different implementation of the NAL-NL1 prescription targets or to differences in the calibration and analysis of the probe microphone systems.

9.5.2 Terminology and Applications

The most commonly used real-ear measurement procedures are discussed in Sects. 9.5.2.1–9.5.2.7. Standardized definitions and measurement procedures are provided in ANSI S4.46-2013.

9.5.2.1 Real-Ear Unaided Response

The real-ear unaided response (REUR) is the sound level, as a function of frequency, at a specified point in the unoccluded ear canal for a specified sound-field level. This can also be expressed as real-ear unaided gain (REUG; the unaided sound level in the ear canal subtracted from the sound level at a reference point near the external ear). For example, a patient who has a 77-dB SPL REUR at 3,000 Hz for a 60-dB SPL input signal would have a 17-dB REUG. The REUR reflects the resonance characteristics of the ear canal and the concha. The most common clinical use of the REUR is to serve as a reference value for the calculation of insertion gain (defined in Sect. 9.5.2.5). It is also possible to calculate insertion gain using an average REUG stored in the equipment. If ear canal sound level targets are used for verification, the aided response is the measure of interest, and there is no reason to measure the REUR.

9.5.2.2 Real-Ear Occluded Response

The real-ear occluded response (REOR) is the sound level, as a function of frequency, at a specified point in the ear canal for a specified sound-field level with the hearing aid (and its acoustic coupling) in place and turned off. This can also be expressed as real-ear occluded gain (REOG), which is the difference between the occluded and nonoccluded ear canal sound level. The purpose of measuring the REOR is to determine the effect of the hearing aid and its earmold coupling on the input to the ear. The difference between the REUR and the REOR is the insertion loss. For some frequencies, this can be close to zero, especially for an "open fitting," but for hearing aid fittings using an occluding earmold, the REOR will often fall below the REUR because of the loss of the natural ear canal resonance and the attenuation properties of the earmold/hearing aid. The REOR and the calculated REOG are not directly used in the fitting process.

9.5.2.3 Real-Ear Aided Response

The real-ear aided response (REAR) is the sound level, as a function of frequency, at a specified measurement point in the ear canal for a specified sound-field level with the hearing aid (and its acoustic coupling) in place and turned on. If the input level has been subtracted, this then becomes the real-ear aided gain (REAG). In the past, REAR was often referred to as the in situ response or the in situ gain. Historically, the REAR was measured to calculate insertion gain. Increasingly, prescription target values are provided as REAR, and, in this case, the REAR becomes the verification metric.

9.5.2.4 Real-Ear Saturation Response

The real-ear saturation response (RESR) is the lowest level, as a function of frequency, at a specified measurement point in the ear canal for a specified sound-field level with the hearing aid in place and turned on, at which the hearing aid operates at its maximum output level. This term is not recommended in ANSI S3.46-2013, the preferred term being REAR85 or REAR90 (i.e., the REAR measured with an input level of 85 or 90 dB SPL, which is assumed to cause most hearing aids to reach their highest possible output level). In the clinic, we are not concerned with the highest *possible* output of the hearing aid but rather with the maximum output that will be produced given the current hearing aid settings. The most common clinical use of the RESR, therefore, is to ensure that the maximum output of the hearing aid is close to the prescription target and does not cause loudness discomfort. Although it can be measured using real speech, the RESR (or REAR85) is often measured using a swept tone, as this drives the hearing aid to a higher output level in each channel, and makes it easier to determine the programmed maximum power output (MPO) at all frequencies.

9.5.2.5 Real-Ear Insertion Gain

The real-ear insertion gain (REIG) is the difference, as a function of frequency, between the REAR and the REUR measured at the same reference point and under the same sound-field conditions. It is a derived measure. The clinical use of the REIG is to verify that the hearing aid meets the prescribed insertion gain.

9.5.2.6 Real-Ear to Coupler Difference

The real ear-to-coupler difference (RECD) is the difference in decibels, as a function of frequency, between the sound level produced near the tympanic membrane in an occluded ear canal by a coupled sound source having high acoustic impedance and the sound level produced in the HA-1 configuration of the 2-cc coupler by the same coupled sound source. If the coupled sound source is the same for the ear canal and the coupler and if the acoustic impedance is higher than that of the ear canal and the coupler, the RECD is independent of the coupled sound source; otherwise, the difference in level between the ear and the coupler depends on the sound source (Munro and Salisbury 2002; Munro and Toal 2005). The generic term for this acoustic transfer function, used in ANSI S3.46-2013 (2013), is the ear-to-coupler level difference (ECLD). Information about potential sources of error in the measurement of the RECD is provided in Annex C of ANSI S3.46-2013 (2013).

There are three clinical applications of the RECD: (1) estimating the sound level produced near the tympanic membrane by a hearing aid from the sound level it produces in a 2-cc coupler (to predict real-ear response when the hearing aid response has been measured in the 2-cc coupler); (2) adjusting hearing threshold levels, measured with an insert earphone and standard ear tip, to those that would have been obtained had the acoustic impedance of the individual ear been that of an average adult ear (e.g., predicting the hearing level of an infant when their external ear assumes the same dimensions as for a typical adult); and (3) correcting hearing level, measured with an insert earphone and a customized earmold, for differences between the earmold and that of the standard ear tip. The RECD is especially relevant when prescribing and fitting hearing aids to infants and children (see Sect. 9.5.3).

9.5.2.7 Real Ear-to-Dial Difference

The real ear-to-dial difference (REDD) is the difference in decibels, as a function of frequency, between the sound level produced near the tympanic membrane by an audiometer and the hearing level indicated by the audiometer. The REDD is used to convert audiometric thresholds (and LDLs) in decibels HL to ear canal sound levels in decibels SPL, as displayed on an SPLogram.

9.5.3 Clinical Protocol

The type of real-ear measure used partly depends on the prescriptive approach. Traditionally, most clinicians have used REIG, primarily because early prescription methods provided gain-frequency response targets. More recently, prescription methods have also provided REAR targets or REAG targets. Both approaches are currently in use, although there are country-specific preferences. In the United States, for example, more than 80 % of hearing aid dispensers now report using

REAR as their primary method for verifying the match to the prescription target (Mueller and Picou 2010). In the United Kingdom, the primary approach is to use REIG with adults and REAR with children.

Regardless of whether the REIG or REAR is used, the typical clinical protocol for verification of hearing aids in cooperative adults takes around 20–30 min to complete for a bilateral fitting and consists of two stages: preparation for verification followed by the real-ear measurements and adjustment where necessary.

In the preparation stage, the audiometric information is entered into the hearing aid fitting software and the real-ear measurement system. In some integrated systems, the audiometric data are automatically transferred from the fitting software to the real-ear measurement system. Althugh RECD transfer functions are not commonly measured for adults, if these are available, they should be entered into the fitting software because this individualizes the conversion from hearing level to ear canal sound level. The probe microphone is then calibrated and the prescription method to be used in the fitting process is selected in the software and programmed into the hearing aids. The patient is seated in front of the loudspeaker, the probe assembly is placed on the ear, and the probe tube inserted into the ear canal. The hearing aid is then placed on the ear. Many probe systems have a right and left probe assembly, so both ears can be prepared for testing in advance.

During verification, the patient is required to sit still and be quiet. Verification of the gain-frequency response typically commences with measurement using a 50-dB SPL speech input (although some audiologists start with a 65-dB SPL speech input). Adjustments are made so that the ear canal sound level is within 3–5 dB of the prescription target for frequencies up to at least 4,000 Hz. These adjustments are made for each channel in the hearing aid. Once a close match to target has been obtained, the gain-frequency response is measured with an 80-dB SPL speech input. The compression threshold or/and ratio is then adjusted until a match to target has been obtained. Some manufacturers have simplified this adjustment by simply having a tab labeled “Gain for Loud” or “Gain for 80 dB” in the fitting software.

The next step is to measure the gain/frequency response with a speech input level of 65 dB SPL. Because 65 dB SPL is the midpoint between 50 and 80 dB SPL and most compression systems use “straight” compression, the response at 65 dB SPL should be very close to the target values. If required, the compression characteristics are adjusted to improve the match to the target values. The final stage of clinical verification of gain-frequency response is to repeat the measurements with input levels of 50, 65, and 80 dB SPL, to view the family of frequency response curves, and to check that the programming alterations for one input level did not have a significant effect on the responses for any other input level.

Clinical verification of the maximum power output of the hearing aid is performed with a swept pure tone at 85 dB SPL (see RESR85, described in Sect. 9.5.2.4). If the output exceeds the patient’s LDL, frequency-specific adjustments can be made by adjusting the maximum output in individual channels. It is recommended that behavioral judgments of loudness accompany this measure.

Real-ear verification requires the passive cooperation of the patient and this can be a challenge for infants and young children. For this reason, an alternative method of verification is required for this population. One approach is to simulate the real ear by measuring hearing aid performance in a 2-cc coupler and then adding the RECD, taking into account microphone location effects (Bagatto and Scollie 2011). Adjustments to the hearing aid response can then be carried out in a 2-cc coupler and do not require the involvement of the patient.

Numerous studies have revealed the extent to which the acoustic properties of the ear canal (and the RECD) change during the first few years of life. As a child grows, the ear canal increases in volume and the RECD becomes smaller, especially at higher frequencies. However, RECDs vary markedly across individuals within the same age range (Feigin et al. 1989; Bagatto et al. 2002). This makes it difficult to predict what the RECD will be for a given ear. For this reason, it is advisable to measure the RECD for each infant. Several studies have demonstrated the validity of using an individually measured RECD to derive the real-ear sound level based on responses measured in a 2-cc coupler (e.g., Seewald et al. 1999; Munro and Hatton 2000).

Although the time required to perform an RECD measurement is typically on the order of 1–2 min, the limited cooperation of some participants means that it may not be possible to complete the procedure for both ears. The difference between the RECD values obtained from the right and left ears is typically only a few decibels (Munro and Buttfield 2005; Munro and Howlin 2010), so if measurements can be completed for one ear, the RECD for the other ear can be predicted with reasonable accuracy. If the measurement cannot be completed for either ear, then it is necessary to resort to age-appropriate correction factors. Following simulated real-ear verification, the hearing aids can then be fitted to the infant.

9.5.4 Verification of Additional Hearing Aid Features

There are a number of hearing aid features (e.g., frequency lowering, directional microphone, digital noise reduction, and automatic feedback reduction) that the audiologist may wish to verify. Although precise characterization of these features may not be possible in the clinical setting, it is possible to use real-ear measurement systems to perform a few simple measurements that indicate if the features are broadly working as expected.

9.5.4.1 Frequency Lowering

The general principle regarding frequency lowering is to shift high-frequency components of speech signals that are inaudible to the patient into a lower frequency region where there is better hearing and a greater chance of audibility. A 6,000-Hz signal, for example, might be moved to 3,000 Hz, either through frequency compression or a uniform downward shift of a block of frequencies. The audiologist needs to decide if the lowered signal is now audible. Conventional real-ear procedures can show if the high-frequency signals have been lowered (there will be a reduction in output for these frequencies), but they do not clearly show the status of the signals in the lower frequency region where they were placed, as they are mixed with the original speech signals from this same region. An exception is hearing aids that incorporate conditional transposition/lowering, which is activated only when there is a high ratio of high- to low-frequency energy. With these devices, the frequency-lowered signal should be easily measured. Some real-ear measurement systems have test signals that allow the audiologist to examine the lowered signals and compare the level of these signals to the patient's hearing thresholds.

9.5.4.2 Directional Microphones

There are three reasons why the audiologist might want to assess directional microphones on the real ear. The first is to confirm that the directional effect is what would be expected from the product. The second is to assess the effect of positioning the hearing aid on the head. This can be especially critical with mini-BTEs, where positioning the hearing aid to be comfortable or unobtrusive might work in opposition to good directivity (the on-head microphone port alignment is critical). The third reason is to assess the effect of the degree of occlusion produced by the earmold on the directional processing. The directivity index (DI) of an open/unoccluded fitting might be considerably less than that of an occluded fitting.

The general protocol for checking a directional microphone is to measure the real-ear output for a signal presented at 0° azimuth and at 180° azimuth (with the hearing aid on "fixed directional" to ensure directional processing is being used). The output should be considerably lower for the 180° condition. Because the 0° and 180° signals are not presented simultaneously, the amplitude compression in the hearing aid will cause the differences between the front and back signals to be smaller than would be obtained in a real-world listening situation. The "true" front-to-back ratio (FBR) of the hearing aid is the difference between the two measures (0° and 180°) multiplied by the compression ratio, provided that the signal level is above the compression threshold at the frequency of interest.

When conducting real-ear measurements, directionality has typically been expressed as the FBR. Recent research, however, suggests that it might be better to use the front-to-side ratio (FSR). Wu and Bentler (2012) found that the FSR predicted the DI and performance on a speech-in-noise task more accurately than did the FBR.

9.5.4.3 Digital Noise Reduction

Digital noise reduction (DNR) algorithms vary considerably across hearing aid manufacturers. A medium setting might mean an 8-dB noise reduction for one manufacturer but 3 dB for another. The speed of operation of the DNR algorithms also varies considerably across manufacturers. This can be observed when conducting a real-ear measure and can be beneficial in patient counseling. To assess DNR, the hearing aid is set to the normal user setting, and a real-ear measure is made with the DNR disabled and then enabled. Most real-ear measurement systems have one or more noise signals that can be used for this purpose. As with directional microphones, open fittings have a substantial impact on the degree of DNR at lower frequencies; for a truly open fitting, the noise reduction will be zero at lower frequencies, regardless of the strength of the setting. This is important for patient counseling, as the low-frequency components in background sounds are often the most noticeable. The main interest in assessing DNR at the time of fitting is to examine the effects of the acoustic coupling on the DNR.

9.5.4.4 Automatic Feedback Reduction

Almost all contemporary hearing aids have sophisticated acoustic feedback reduction systems. However, the effectiveness of these systems can vary by 10 dB or more among manufacturers or among different models from the same manufacturers. The process for determining the effectiveness of the feedback system is straightforward. After programming the hearing aid to a validated prescriptive method, a real-ear response is measured at this setting with a 65-dB SPL input. This serves as the baseline. The feedback reduction system is disabled and the overall gain is increased gradually until audible feedback just occurs; again, a real-ear response is measured. Next, the feedback reduction system is enabled and overall gain is gradually increased until audible feedback just occurs. Another real-ear response with an input of 65 dB SPL is recorded. The difference between these latter two responses is what is commonly referred to as "added gain before feedback", that is, the improvement provided by the feedback suppression circuitry.

9.5.5 Real-Ear Measurements in Clinical Practice

When using real-ear measures to fine tune a hearing aid to prescriptive target values, it is reasonable to ask "how close is close enough?" In research studies, it is common to have the goal of programming the hearing aids within 2–3 dB of prescriptive target (e.g., Moore et al. 2001). Cox and Alexander (1990) reported that differences among prescriptions were preserved if the RMS error for the match to target was 5 dB or less. Tolerances of 3–5 dB, therefore, are commonly allowed in clinical practice. However, there are several factors to be considered. For example, it is desirable that the overall frequency-gain characteristic should have the prescribed shape. A fitting that is 5 dB below target at all frequencies is preferred to one that is 5 dB above at some frequencies and 5 dB below at others, even though the overall fitting error might be the same. Other factors that might affect the fitting tolerances include

- Does the patient have a volume control for making gain changes? If not, the accuracy of the fitting is more critical.
- Is the patient using trainable hearing aids? Using data logging, many modern hearing aids allow users to train the gain and frequency response of their hearing aids to their preferred values. The hearing aids "remember" what adjustments are made for different listening environments (for a review, see Mueller 2014a). In this case, patient preferences will play a part, and the fitting may not need to be as accurate on the day of the fitting (although the overall shape of the frequency-gain characteristic remains important).
- The amount of gain preferred by the patient may gradually increase over time. To allow for this, many hearing aids automatically provide a gradual increase in gain over time. If so, it is acceptable for the initial output levels to be somewhat below the target values.
- Is the programmed gain "acceptable" to the patient? That is, is the patient willing to wear the hearing aid regularly at the given gain settings?

Does clinical verification and matching to validated prescription targets lead to a better outcome for the patient? To assess this, Keidser and Alamudi (2013) fitted 26 hearing-impaired individuals (experienced hearing aid users) with new trainable hearing aids, which were initially programmed to NAL-NL2 targets. The training algorithm was a revised version of SoundLearning (Chalupper et al. 2009), which enables independent training of gain below the compression threshold and of the compression ratios in four frequency bands by relating selected gain to the input level in each band. This means that the gain-frequency response shape is also trained. Following 3 weeks of training, they examined the new trained settings for both low and high frequencies for six different listening situations (the training was situation specific). The participants tended to reduce the gain relative to NAL-NL2 targets but only by a minimal amount. For example, for the speech in quiet condition, the average gain reduction at high frequencies was 1.5 dB (95 % range = 0 to −4 dB), while for the speech-in-noise condition, it was 2 dB (95 % range = 0.5 to −4.5 dB). The trained gains for low-frequency sounds for these listening conditions were even closer to the NAL-NL2 targets.

Perhaps even more compelling data come from another study using input-specific and situation-specific learning hearing aids conducted by Palmer (2012). The participants were 36 new users of hearing aids. One group of 18 was fitted using NAL-NL1 targets, used this gain prescription for 1 month, and then trained the hearing aids for the following month. A second group of 18 was also fitted using NAL-NL1 targets but started training the aids immediately and trained them for 2 months. Importantly, the gains could be trained up or down by 16 dB, providing ample opportunity for participants to achieve their preferred loudness levels. After 2 months of hearing aid use, both groups ended up using gains very close to those prescribed by NAL-NL1. Palmer reported that the SII for soft speech was reduced by a modest 2 % for the first group, and by 4 % for the second group.

As mentioned in Sect. 9.4.4, Abrams et al. (2012) compared the initial gains provided by the manufacturers' fitting software and gains obtained after real-ear verification and adjustment to match NAL-NL1 target values, using 22 participants. The adjusted gains were somewhat below NAL-NL1 targets but were significantly closer to the prescribed targets than the gains obtained with the manufacturers' initial software. The participants used the hearing aids with the two different fittings in the real world, and following each trial period, completed a validated self-report questionnaire called the Abbreviated Profile of Hearing Aid Benefit (APHAB; Cox and Alexander 1995). After using both fitting methods, the participants selected their preferred fitting. The APHAB mean results showed significant overall preferences for the verified/adjusted fitting for speech in quiet, in reverberation, and listening in background noise. Of the 22 participants, 17 selected the verified fitting as their preferred fitting. For several of the participants, the verified fit and the initial fit were not significantly different. Of the 13 participants for whom the error was significantly larger for the initial fit than for the verified/adjusted fit, 11 preferred the verified fit at the conclusion of the study. These data suggest that, on average, fittings approaching the NAL-NL1 targets provided more real-world benefit than the initial fits, which typically led to lower gains than the NAL-NL1 targets.

Given the preciseness of real-ear measurements, their ease of use, the importance of aided audibility, and the impact that well-fitted hearing aids have on a patient's life, one might assume that real-ear measures are common in clinical practice. However, numerous studies have suggested that this is not the case (Kochkin et al. 2010; Mueller and Picou 2010; Mueller 2014b). The audiologist must first accept that the typical patient will have a better outcome if fitted to a validated prescriptive target. The audiologist also needs to be aware that using the initial settings selected by the manufacturers' software does not ensure a good match to the prescription target values (Mueller 2014b). For example, Aazh and Moore (2007) and Aazh et al. (2012b) have shown that, at least for some manufacturers, the real-ear output will miss the target values at one or more frequencies by 10 dB or more for over 60 % of patients when this approach is used. Once the professional has acquired this basic knowledge and established these beliefs, then conducting real-ear measurements for all patients is simply a matter of professional integrity.

9.6 Summary

The general goal of providing amplification is to improve functional auditory capacity and restore good communication skills. Amplification should restore the audibility of soft sounds, provide improved intelligibility of speech at conversational listening levels, and ensure that intense sounds are not amplified to an uncomfortably loud level. There are several prescription methods that provide frequency-specific target values for soft, conversational, and intense sounds. Despite differences in the target values, no validated prescription method has been clearly shown to be superior to any of the other methods in terms of patient benefit (e.g., greater satisfaction, less residual disability). However, clinical studies have clearly shown that when a well-researched prescriptive approach is used and appropriate gain is delivered across frequencies, speech intelligibility is enhanced, and there is improved patient benefit and satisfaction. There is also irrefutable evidence that the audiologist can improve the match to the prescription target values using a probe microphone placed within the patient's ear canal. As a result, carefully conducted verification is an essential component to long-term success with amplification. The most recent generation of prescription methods provide a degree of personalization to the target values beyond that associated with hearing threshold levels. However, there is an urgent clinical need to address the wide range of clinical outcomes that occur in hearing aid users with apparently similar characteristics.

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Chapter 10
Hearing Aid Validation

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Abstract Validation provides quality assurance that a hearing aid wearer's needs are being met—that the solution meets not only their technical requirements (i.e., verification) but also their requirements for everyday communication. In the past 50 years, there have been repeated calls for better measures of hearing aid performance, with a general shift in validation toward the self-report of hearing, communication, and well-being through questionnaires. This chapter looks at these measures, examining the domains of hearing aid validation and how despite the growth in number of questions—a total of more than 1,000 questions on hearing aids—the domains have evolved only slightly. The chapter then considers the ways in which a fundamental domain, "benefit," is calculated. A large data set shows how different forms of benefit can lead to different systematic interpretations. While most objective measures for hearing aids are by definition verifications, the chapter discusses those objective measurements that approach validation by attempting to mimic aspects of everyday communication. The issues raised by these myriad forms of validation suggest that a viable measure of hearing aid benefit must incorporate measures of expectations and burdens for listener-specific conditions.

Keywords Benefit • Hearing aids • Questionnaires • Self-report • Speech-in-noise • Validation

10.1 Introduction

Validation, as opposed to verification, provides quality assurance that listeners' needs in terms of hearing aids are being met—that the solution meets not only their technical requirements (which is verification) but also their requirements in

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everyday life. For example, at a hearing aid fitting, clinical protocols are used to verify that the targets for real-ear insertion gains are reasonably well met; see Munro and Mueller, Chap. 9. But these clinical protocols do not assess whether those gains meet the needs of the hearing aid user, if they alleviate disability and handicap, or if they relieve restrictions or limitations due to hearing loss. The true goal of validation, as noted by Carhart in 1965, is “to evaluate a patient’s everyday difficulties in hearing and to assess the practical significance of therapeutic and rehabilitative procedures for him [sic]. Here, the goal is to diagnose the social efficiency of his audition” (p. 259). The domains of validation are the measurement of benefit and the “practical significance” of the hearing aid fitting, the gains in “social efficiency,” and/or decreases in everyday difficulties.

In the past 50 years, there have been regular and repeated calls for better measures of hearing aid performance, especially with regard to patient benefit (e.g., Hawkins 1985; Gatehouse 1993; Danermark et al. 2010). There has been a general shift in validation since the mid-1980s toward the patient’s (self) report of hearing, communication, and well-being through questionnaires. A quest for better self-report validation has propelled the continued invention and modification of hearing aid questionnaires. Previous reviews have looked at the rigor—the statistical validity and repeatability—of the questionnaires (e.g., Mendel 2007) and how questions fit into current domains of health status (e.g., Granberg et al. 2014). Other reviews have looked at the clinical application(s) of prominent hearing questionnaires (Bentler and Kramer 2000; Cox 2005). This chapter looks at the domains of hearing aid validation, the questions, and if they have evolved over time. To limit the myriad forms of validation, the chapter considers only subjective reports for aided adults.

After surveying the plethora of self-report instruments that have been developed for measuring validation, the chapter focuses on the primary concern, “benefit”, but even this has different interpretations. The chapter considers examples of hearing aid benefit determined from a large data set obtained using the Glasgow Hearing Aid Benefit Profile (GHABP) questionnaire (Gatehouse 1999), and then uses these data to determine how benefit relates to satisfaction, and how balancing the two may lead to success. The discussion then turns to attempts at objective validation. By definition, validation usually precludes the use of objective measurement; most objective measures of hearing aid performance are considered verification: ensuring that the hearing aid performs as expected. The objective challenge is to mimic everyday listening, to measure communication as it occurs in complex situations (e.g., Hafter et al. 2012) and continuously (e.g., MacPherson and Akeroyd 2014). Finally, there is a summary of the issues relating to how Carhart’s “social efficiency” can best be measured.

10.2 Subjective Validation

Table 10.1 lists the self-report questionnaires since 1965 that include at least one question explicitly about hearing aids. It is a subset of a larger database of questionnaires on hearing that have been collated at the Institute of Hearing Research,

Table 10.1 List of questionnaires that include at least one question about hearing aids, ordered by year published

Year	Questionnaire	Author(s)	Abbrev.	Q(all)	Q(HA)
1965	**Hearing ability survey**	**Schein et al.**	**HAS**	**53**	**24**
1970	Philadelphia scale	Schein et al.		12	1
1970	Washington scale	Schein et al.		9	1
1977	**Feasibility scale for predicting hearing aid use**	**Rupp et al.**	**FSPHAU**	**11**	**11**
1978	Denver scale of communication function for senior citizens living in retirement centers	Zarnoch et al.	DSSC	7	2
1980	(Hearing screen test for the elderly)	Manzella and Taigman	–	6	1
1980	**Hearing problem inventory (Atlanta)**	**Hutton et al.**	**HPI**	**51**	**10**
1983	National study of hearing	Davis	NSH	21	2
1984	**Hearing aid performance inventory**	**Walden et al.**	**HAPI**	**64**	**64**
1986	**Binaural hearing aid questionnaire**	**Chung et al.**	**BHAQ**	**33**	**33**
1988	**Negative reactions to hearing aids**	**Surr and Hawkins**	**NRHA**	**20**	**20**
1989	**Attitudes towards loss of hearing questionnaire**	**Brooks[a]**	**ALHQ**	**24**	**6**
1990	**Expectations checklist**	**Seyfried[b]**	**EC**	**13**	**13**
1990	**Profile of hearing aid performance**	**Cox and Gilmore**	**PHAP**	**66**	**66**
1991	**Intelligibility rating improvement scale**	**Cox et al.**	**IRIS**	**66**	**66**
1991	**Profile of hearing aid benefit**	**Cox et al.**	**PHAB**	**66**	**66**
1995	(Disabilities and handicaps associated with impaired auditory localization)	Noble et al.	–	38	2
1995	Communication self-assessment scale inventory for deaf adults	Kaplan et al.	CSDA	125	2
1996	**Hearing attitudes in rehabilitation questionnaire**	**Hallam and Brooks**	**HARQ**	**40**	**16**
1997	**Client-oriented scale of improvement**	**Dillon et al.**	**COSI**	**10**	**10**
1997	Communication scale for older adults	Kaplan et al.	CSOA	72	2
1997	**Hearing aid satisfaction survey**	**Kochkin**	**HASS**	**39**	**39**
1999	**Glasgow hearing aid benefit profile**	**Gatehouse**	**GHABP**	**24**	**16**
1999	**Hearing aid users questionnaire**	**Dillon et al.**	**HAUQ**	**22**	**22**
1999	**Hearing disability and aid benefit inventory**	**Gatehouse**	**HDABI**	**126**	**72**
1999	**Profile of aided loudness**	**Palmer et al.**	**PAL**	**24**	**24**
1999	**Satisfaction with amplification in daily life**	**Cox et al.**	**SADL**	**15**	**15**
2000	**Expected consequences of hearing-aid ownership**	**Cox and Alexander**	**ECHO**	**15**	**15**

(continued)

Table 10.1 (continued)

Year	Questionnaire	Author(s)	Abbrev.	Q(all)	Q(HA)
2000	**International outcome inventory for hearing aids**	**Cox et al.**	**IOI-HA**	**7**	**7**
2001	**Hearing satisfaction scale for hearing aids**	**Stewart et al.**	**HSS**	**15**	**15**
2002	**Glasgow hearing aid difference profile**	**McDermott et al.**	**GHADP**	**24**	**24**
2004	**Hearing aid performance questionnaire**	**Vestergaard**	**HAPQ**	**18**	**18**
2004	**Speech, spatial, & quality of hearing questionnaire**	**Gatehouse and Noble**	**SSQ-B/C**	**49**	**49**
2005	**(Self-assessment Inventory)**	**Meister et al.**	**–**	**33**	**18**
2005	**Attitudes towards loss of hearing questionnaire v2.1**	**Saunders et al.**	**ALHQ v2.1**	**22**	**6**
2005	**Client satisfaction survey**	**Uriarte et al.**	**CSS**	**16**	**16**
2005	**Effectiveness of auditory rehabilitation scale**	**Yueh et al.**	**EAR**	**13**	**11**
2007	**(Early screening for hearing disability)**	**Davis et al.**	**HTA2007**	**141**	**84**
2009	**Audiological rehabilitation—clinical global impression**	**Öberg et al.**	**AR-CGI**	**16**	**13**
2009[c]	**Device-oriented subjective outcome scale**	**Cox et al.**	**DOSO**	**40**	**40**
2009	**Environmental sounds questionnaire**	**Blamey and Martin**	**ESQ**	**36**	**36**
2009	**Self-assessment of communication (Revised)**	**Ivory et al.**	**SAC-Hx**	**12**	**12**
2010	**Intervention questionnaire**	**Laplante-Lévesque et al.**	**IQ**	**52**	**20**
2011	**Bern benefit single-sided deafness questionnaire**	**Kompis et al.**	**BBSS**	**10**	**10**
2011	**Own voice qualities—monaural/binaural user**	**Laugesen et al.**	**OVQ**	**100**	**90**
			Total	**1676**	**1090**

Questionnaires that included more than two questions about hearing aids are in boldface and are discussed in the text. Abbreviated forms without modified questions are not included in the list but are discussed in the text

[a]Questionnaire published in Saunders and Cienkowski (1996)

[b]Questionnaire published in Bentler et al. (1993)

[c]Questionnaire available in 2009, published in 2014

beginning with the archives of Stuart Gatehouse. Inclusion in this database was limited only by the availability of the questions and by the questionnaire being targeted to adults; pediatric/parental self-report instruments as well as instruments devoted to bone-anchored hearing aids or cochlear implants have been omitted. The full database has 181 questionnaires. There are 138 hearing questionnaires that are available in English, and of those, 109 are questionnaires that were not shortened forms of another questionnaire. The 45 questionnaires that assessed hearing aids or included at least one question explicitly about hearing aids are listed in Table 10.1. The sheer number of questionnaires is clear, as is the fact that there was a steady progression of publications of new questionnaires that at the turn of the millennium grew exponentially. There are more than 1,000 questions on hearing aids in all, although many are not unique.

10.2.1 A History of Hearing Aid Questionnaire Domains

The first hearing questionnaire to involve hearing aids was the Hearing Ability Scale (HAS) devised by Schein et al. (1965). The HAS includes questions on the type of hearing aid being worn, how much it is used ("never/once-in-a-while/most-of-the-time") in a variety of situations (at work, school, church, theater, home, and listening to radio/TV), whether sounds are audible, if speech is understandable both with and without visual information (e.g., lip reading), and if telephone conversation is possible with the hearing aid. Later questionnaires expanded greatly on the situations assessed as well as the assistance provided with and without visual information.

After token questions concerning hearing aid ownership and use in surveys of 1970 (the Philadelphia and Washington scales; National Center for Statistics) and 1971 (the Denver Scale; J. Alpiner, W. Chevrette, G. Glascoe, M. Metz, and B. Olsen, unpublished study, University of Denver, 1971), the first hearing aid specific questionnaire appeared in 1977: the Feasibility Scale for Predicting Hearing Aid Use (FSPHAU; Rupp et al. 1977). The FSPHAU, however, was intended to predict candidacy for future hearing aid(s), not validate hearing aid(s), using a combination of audiometric data and measures of motivation. In a revised form, it was found to have poor test–retest reliability (Chermak and Miller 1988), and there is no evidence that it was ever used again.

Before specific hearing aid validation questionnaires were developed in the 1980s, multiple general hearing questionnaires were applied to hearing aids. The Hearing Handicap Scale (HHS; High et al. 1964) asks 40 questions, split into two balanced parts, on the ability to detect and understand speech in a variety of situations (e.g., "Can you carry on a conversation with one other person when you are on a noisy street corner?") with responses on a continuous scale with five anchors from "never" to "always." Tannahill (1979) used the HHS to measure benefit, asking new hearing aid wearers to complete one-half of the HHS on their hearing without their hearing aid and then the other half on their hearing with the hearing aid(s) 4 weeks after fitting. Tannahill found increases in mean abilities after fitting (i.e., evidence of hearing aid benefit), but the minority of patients who did not continue use of their hearing aids after the trial reported either a deficit—decreased abilities postfitting—or benefits below the mean. Thus, one of the first self-report instruments for validating hearing aids showed both the benefit of amplification and a relationship between benefit and acceptance of amplification.

The Hearing Performance Inventory (HPI; Giolas et al. 1979), a more general and lengthy hearing questionnaire, was used to evaluate hearing aid fittings for those with profound hearing losses (Owens and Fujikawa 1980), showing increased performance for aided versus unaided respondents. The HPI includes six subscales or domains across 158 questions: speech understanding with and without visual cues (e.g., “You are with a male friend or family member in a fairly quiet room. Can you understand him when his voice is loud enough for you and you can see his face?”); sound detection (e.g., “Can you hear an airplane in the sky when others around you can hear it?”); response to auditory failure (e.g., “You are talking with a stranger. When you miss something important that was said, do you ask for it to be repeated?”); social (which reuses speech understanding and auditory failure questions concerning groups; e.g., “You are talking with five or six friends. When you miss something important that was said, do you ask the person talking to repeat it?’); personal (self-esteem and social interactions; e.g., “Does your hearing problem lower your self-confidence?”); and occupational (which repeats speech understanding and auditory failure questions in work-related scenarios; e.g., “You are talking with your employer [foreman, supervisor, etc.] at work. When you miss something important that was said, do you pretend you understood?”). The Hearing Problem Inventory (HProbI; Hutton 1980) includes 10 questions about (1) operating hearing aids (e.g., the frequency of trouble with earmolds), (2) use, (3) satisfaction, and (4) how much help in general the hearing aids provide. The Hearing Handicap Inventory for the Elderly (HHIE; Ventry 1982), a general hearing questionnaire like the HHS and HPI, has also been used to validate hearing aid interventions (e.g., Newman and Weinstein 1988). The HHIE covers the emotional toll (e.g., “Does a hearing problem cause you to feel depressed?”) and social restrictions of hearing loss (e.g., “Does a hearing problem cause you to avoid groups of people?”).

Hearing aid specific questionnaires followed. The Hearing Aid Performance Inventory (HAPI; Walden et al. 1984) applies the situational aspect of the HPI specifically to hearing aid benefit; it assesses how much the hearing aid helps across 64 different situations (e.g., “You are shopping at a large busy department store and are talking with a salesclerk”). The Binaural Hearing Aid Questionnaire (BHAQ; Chung and Stephens 1986) asks questions regarding wearing *two* hearing aids as opposed to one: use, socializing, ease of listening, and locating sounds (e.g., “When you are at meetings, church, pictures, or theaters, do you find listening easier using….one hearing aid, two hearing aids, no difference?”). The Hearing Aid Users Questionnaire (HAUQ; Forster and Tomlin 1988; cited in Dillon et al. 1997) asks about operating the hearing aid, use, benefit in six situations, satisfaction and the quality of service (e.g., “Do you have any difficulty adjusting the controls of the hearing aid?”). Two questionnaires followed that probed the stigma—the psychological burden—of hearing aids: the Negative Reactions to Hearing Aids questionnaire (NRHA; Surr and

Hawkins 1988) and the Attitudes toward Loss of Hearing Questionnaire (ALHQ; Brooks 1989). The NRHA questionnaire asks 22 questions on how hearing aids are perceived by the respondent and others in terms of benefit, self-image (e.g., “People will notice the hearing loss more when I wear the hearing aid”) and sociability [e.g., “People will be (are) reluctant to talk to me when they see the hearing aid”]. The ALHQ includes four questions on attitudes toward and image of wearing hearing aids (e.g., “Are you concerned about being seen wearing hearing aids?”).

Despite the breadth of domains established for subjectively assessing hearing aids in these questionnaires up to 1990, the development of new questionnaires continued, bolstered at least in part by greater weight being given to subjective report (Cox 1999). But with occasional exception, the domains and questions were mostly repeated from previous questionnaires, although with continuing focus on psychosocial or speech benefit. For instance, the Profile of Hearing Aid Performance (PHAP; Cox and Gilmore 1990) uses some of the same questions as the HAPI, focusing on the frequency of difficulties (never to always) across numerous situations (e.g., “I can understand conversations even when several people are talking”). Immediately following the PHAP were the Profile of Hearing Aid Benefit (PHAB; Cox et al. 1991a) and the Intelligibility Rating Improvement Scale (IRIS; Cox et al. 1991b), both using the same questions as the PHAP but with responses based on the frequency of difficulties and percentage of speech understood, respectively, in those situations. The use of a difference as opposed to absolute scale to measure benefit is discussed in Sect. 10.3.

Abbreviated forms of previous questionnaires proliferated through the remainder of the 1990s. There were two shortened versions of the HAPI (Schum 1992; Dillon 1994). There were abbreviated versions of the PHAP (Purdy and Jerram 1998) and PHAB (APHAB; Cox and Alexander 1995). Although similar to the PHAB, Gatehouse took a slightly different approach to validating performance in everyday environments with the Hearing Disability and Aid Benefit Inventory (HDABI; Gatehouse 1999), by inquiring about hearing aid use, satisfaction, and benefit in each of 18 situations (e.g., “Listening to the television on your own: In this situation with your hearing aid how much difficulty do you have?”). This was truncated to four situations involving speech—conversation in quiet, television listening, noisy street conversation, and group conversation. This truncated form is the Glasgow Hearing Aid Benefit Profile (GHABP; Gatehouse 1999). The GHABP asks about unaided disability and handicap in terms of annoyance and, when aided, benefit, duration of use, aided (residual) disability, and satisfaction for each of the four situations. It also allows for the six questions to be applied to user-defined situations (see the COSI; Sect. 10.2.2). A slightly modified version of the GHABP, the Glasgow Hearing Aid Difference Profile (GHADP; McDermott et al. 2002) was later created to allow comparison of new versus current hearing aids with the same six questions about the same four situations.

Akin to the psychosocial NRHA questionnaire and ALHQ instruments, the Expectations Checklist (Seyfried 1990; cited in Bentler et al. 1993) probes attitudes toward hearing aids across the domains of the more performance-centered questionnaires: hearing aid operation, use, satisfaction, and speech benefits as well as the expected effects on emotional and social well-being (e.g., “I will feel self-conscious when I am using my hearing aid[s]"). The Hearing Attitudes in Rehabilitation Questionnaire (HARQ; Hallam and Brooks 1996) asks about expectations in relation to hearing aids in addition to general attitudes toward hearing loss (e.g., "I would expect to get used to using a hearing aid in a matter of days"). The Hearing Aid Needs Assessment (HANA; 1999) similarly asks about expectations in relation to hearing aid help in 10 specific scenarios (cf. the HAPI; e.g., "You are at church listening to a sermon and sitting in the back pew: How much help do you expect the hearing aid to provide?").

The Satisfaction with Amplification in Daily Life questionnaire (SADL; Cox and Alexander 1999) asks 15 questions on a range of topics, from general benefit (e.g., "Do you think your hearing aid[s] is worth the trouble?") to the competence of the hearing aid provider (e.g., "How competent was the person who provided you with your hearing aid[s]?"). The Profile of Aided Loudness (PAL; Mueller and Palmer 1998) asks about perceived loudness and satisfaction with loudness ("is the loudness appropriate?") for 12 everyday sounds. The PAL provides an instance of a self-report instrument for assessing what is considered clinical verification: are the gains delivered by the hearing aid(s) appropriate? The 12 everyday sounds of the PAL were modified and expanded to 18 by Blamey and Martin for their Environmental Sounds Questionnaire (2009; e.g., "*For each of the following sounds: Running water, such as a toilet or shower*"). A much more systematic and commercially focused questionnaire is the Hearing Aid Satisfaction Survey (HASS; Kochkin 1997), developed as part of the "MarkeTrak" industry surveys used to provide information for policy, industry, and healthcare provision. The HASS asks 34 questions about satisfaction with hearing aids, ranging from overall satisfaction to satisfaction in specific situations. It includes five additional questions covering willingness to buy, use, and quality of life.

The International Outcome Inventory for Hearing Aids (IOI-HA; Cox et al. 2000) was developed with a different approach from that used for previous questionnaires. A consortium of experts was assembled, the domains important for hearing aid validation were defined, and a single question for each of those domains was developed. Six of the seven resulting domains/questions [use, benefit, "residual activity limitation" (similar to aided disability), satisfaction, "residual participation restriction" (similar to aided handicap), and impact on others] had previously appeared in a variety of questionnaires. The last domain/question simply asks if the hearing aid(s) have improved the quality—"enjoyment"—of life.

Despite its consortium approach, and the considerable expertise that went into the IOI-HA as an attempt to coalesce all hearing aid questionnaires into a standard, the growth of other questionnaires continued. Shortly after the IOI-HA was published, Meister et al. (2001) opted for a conjoint analysis (a regression analysis in which trade-offs/preferences are weighted to determine question rank) instead of a consortium approach and then developed (Meister et al. 2005) a questionnaire based on ten domains, including five hearing aid domains: benefit, satisfaction, usage, importance, and expectation. Each domain was applied to general situations in a single question, except for benefit, which had two questions. A factor analysis indicated that the ten domains could be collapsed into four: restriction (e.g., unaided difficulty), hearing aid benefit, usage, and situation-related factors (e.g., importance and expectation). Vestergaard (2004) developed the Hearing Aid Performance Questionnaire (HAPQ), which asks about how good/bad the device(s) are at keeping sounds natural and comfortable (e.g., "How good or bad are your hearing aids at dealing with the following circumstances: Listening when the volume of music is turned up high?") and how much they help with speech understanding (e.g., "How good or bad are your hearing aids at dealing with the following circumstances: Making speech as understandable as possible"). The Device-Oriented Subjective Outcome scale (DOSO; Cox et al. available online 2009, published 2014) also asks questions from the perspective of what the hearing aid does for the user, based on the finding that personality influenced responses (Cox et al. 2007), about operating the devices, benefits, and enjoyment of life (e.g., "How good are your hearing aids at…Improving enjoyment of everyday activities?"; n.b., the DOSO, scale IOI-HA, and HASS are the only questionnaires that assess enjoyment of life). The Client Satisfaction Survey (CSS; Uriarte et al. 2005) asks very similar questions to the HAS that was published 40 years earlier. The Effectiveness of Auditory Rehabilitation scale (EAR; Yueh et al. 2005) asks about the comfort, operation, value (cf. SADL questionnaire), and convenience of the hearing aid.

A very different self-report instrument was the extensive questionnaire used by Davis et al. (2007) as part of the UK Health Technology Assessment (HTA) study of the prevalence of hearing loss and value of screening for hearing loss. The interview portion of the HTA questionnaire has up to 141 questions (the number of questions is determined by previous responses), with 84 questions on hearing aids covering quality of life, benefit, satisfaction, use, value, stigma, expectations, operation, opinion of services, and a priori knowledge (e.g., "How long do you think it would take to get used to a hearing aid?"). This questionnaire is so long that it has no clinical application or experimental practicability for hearing aid validation, although the data from the larger HTA questionnaire, which included both quality of life and economic cost questions, have since been used to construct economic models to show the value of systematic screening for hearing aid referral (Morris et al. 2013).

Another general hearing questionnaire is the Speech, Spatial and Qualities of Hearing questionnaire (SSQ; Gatehouse and Noble 2004). The SSQ includes 14 questions on speech understanding that form the "speech" subscale (e.g., "You are talking with one other person in a quiet, carpeted lounge room. Can you follow what the other person says?"), 17 questions on locating sounds that form the "spatial" subscale (e.g., "In the street, can you tell how far away someone is from the sound of their voice or footsteps?"), and 18 (originally 19) questions on sound discrimination, naturalness, prosody, and effort that form the "qualities" subscale (e.g., "Does your own voice sound natural to you?"). A recent factor analysis of a large data set showed that the subscales could be reduced to 11, 12, and 12 questions, respectively (Akeroyd et al. 2014). While the SSQ questionnaire has provided evidence for benefits of bilateral fitting, especially in the spatial subscale (Noble and Gatehouse 2006), the SSQ questionnaire has been used relatively little for hearing aid validation since. Perhaps this is because it is twice as long as most contemporary questionnaires and some seven times longer than the IOI-HA, although many shortened forms have been described (Noble et al. 2013).

The Self-Assessment of Communication questionnaire (SAC; Schow and Nerbonne 1982) has six questions on speech understanding in general situations (e.g., "Do you experience communication difficulties in situations when speaking with one other person [for example, at home, at work, in a social situation, with a waitress, a store clerk, with a spouse, boss, etc.?"] and four questions on emotional and social difficulties (e.g., "Do you feel that any difficulty with your hearing limits or hampers your personal or social life?"). It was revised by Ivory et al. (2009), as the SAC-Hx, questionnaire to be used, like the APHAB and others, to validate hearing aids by measuring benefit as the difference between pre- and postfitting ratings. The SAC-Hx questionnaire uses a combination of SAC questions and IOI-HA questions, resulting in six questions on benefit—including the same four situations as the GHABP (television, quiet, noisy, and group conversation), a user-defined situation, and environmental sounds—and one question each on social restriction, handicap, limitation, quality of life, use, and satisfaction. These are the seven domains of the IOI-HA, although with five additional questions on benefit. The SAC-Hx questionnaire is of interest to the surveyor of questionnaires as it explicitly cross-references its questions with those of other questionnaires, highlighting the reconfiguration of the same questions and domains over the years.

Further growth has been due to the focus on different perspectives, such as the Clinical Global Impression of Audiologic Rehabilitation (AR-CGI; Öberg et al. 2009), focusing on longitudinal changes (e.g., "Have you changed your attitude to be a hearing aid wearer over time?"), and the Intervention Questionnaire (IQ; Laplante-Lévesque et al. 2010), which continues the tradition of previous attitudinal questionnaires, the NRHA, ALHQ, and Expectations Checklist but asks about how attitudes brought the respondent to the clinic (e.g., "Hearing aids will prevent my hearing problems from affecting me more in the future: How much did this play a part when I decided what to do for my hearing?"). Specific questionnaires continue, such as the Bern Benefit Single-Sided Deafness Questionnaire (BBSS; Kompis et al. 2011) for evaluating bone-anchored hearing aids (see Killion, Van Halteren, Stenfelt, and Warren, Chap. 3) and the Own Voice Qualities questionnaire (OVQ; Laugesen et al. 2011) for exhaustively evaluating the sound of the user's own voice.

It is telling that in their review of self-report hearing and hearing aid measures, Bentler and Kramer (2000) explicitly recommend using existing questionnaires instead of developing new ones. The list in Table 10.1, however, gives cause to doubt that there will be a pause—or even a slowing down—in questionnaire creation despite the inevitable replication of previous questionnaires.

10.2.2 Open-Set Responses or Predefined Choices?

In their discussion of the use of the HAPI, Walden et al. (1984) noted two problems: (1) an "acquiescence" or predisposition to a (positive or negative) response regardless of inquiry and (2) an inability to judge the hearing aid benefit—whether expected or experienced—in a given situation, resulting in a judgment of the difficulty experienced in that situation. The former issue is usually affected by the patient's overall disposition; the issue of personality can potentially be controlled by the inquiry focusing on how the device is performing (e.g., the DOSO questionnaire) as opposed to how the patient is performing with the device. The latter issue is more problematic: whether the questionnaires cover a large number of scenarios (e.g., HAPI, PHAP, SSQ) or describe simple scenarios (e.g., GHABP), the response may be based on the difficulty experienced in that situation and not the help received from the hearing aid. The tendency to substitute an easy question for a more difficult one is illustrated by our long experience in using the GHABP as part of a screening program for potential research participants (Whitmer et al. 2014). Participants often express confusion in considering the seemingly simple situations covered by the GHABP.

The problem of benefit misjudgment can be avoided by having the patient describe situations they have experienced and can comprehend. Using this "open-set" approach, the Client-Oriented Scale of Improvement (COSI; Dillon et al. 1997) prompts for (up to) five specific situations that are important to the individual (cf. Barcham and Stephens 1980; Stephens 1980). Ease of communication is then rated before and after fitting (or change in hearing aid settings) for those specific situations. The benefit derived from the difference between ratings for pre- and postfitting is considered to be more indicative of the decreased difficulty experienced in that situation as well as avoiding the assumption of experience with the hypothetical situations proposed by other questionnaires. Given the push toward customizing the "patient pathway" for the individual, the open-set approach appears to be a good way forward. Eliciting situations of import from the patient or participant, however, can be difficult. For example, the GHABP also allows for user-defined situations of interest, but they are very rarely used. In our laboratory across many years of testing, of all the participants (1,326) given the option of defining additional situations to rate benefit, use, residual difficulty, and satisfaction, only 5.5 % (73) gave any examples.

10.2.3 Specific or Generic Instruments?

All of the questionnaires listed in Table 10.1 are essentially about hearing. This maximizes their relevance in assessing the benefit provided by hearing aids, but it does not allow comparison with the benefits of other medical interventions (e.g., hip surgery). The problem is exacerbated by hearing impairment being neither life-threatening or (except in rare situations) requiring surgery, so basic across-domain criteria for failure/success, such as mortality rates or surgical success rates, are inapplicable. But such comparisons are becoming important in modern healthcare services as they underpin the calculations of cost effectiveness that are increasingly used to decide which treatments to fund.

To allow for broader comparisons, a general health instrument could be useful. But despite having consistency and reliability and measuring general improvements in communication and participation, generic health instruments, such as the World Health Organization's Disability Assessment Scale version 2 (WHO-DAS II; WHO

2001b), are not as sensitive to the improvements produced by hearing aids as hearing-specific questionnaires. McArdle et al. (2005) found that, for a large pool of veterans (initially 380 randomly split into those who immediately received a hearing aid and those who received a hearing aid 2 months later), followed through 1 year of evaluation, the changes in WHO-DAS II scores produced by provision of hearing aids were much smaller than those for the APHAB and HHIE (effect sizes, η^2, for total WHO-DAS II, HHIE, and APHAB scores were 0.20, 0.74, and 2.19, respectively). More importantly, in that study the participants for whom provision was delayed 2 months showed statistically significant prefitting improvements on both the WHO-DAS II and APHAB questionnaires, indicating that those two questionnaires were potentially susceptible to participant acquiescence. Vuorialho et al. (2006) found that one of the standard generic health instruments, the EuroQOL (EuroQOL Group 1990), was not sensitive to the benefits of hearing aids for first-time users, while the shortened HHIE did show benefits. The other main generic instrument for measuring health-related quality of life, the Health Utility Index (Furlong et al. 2001), is also somewhat insensitive to the effects of hearing impairment and its management by hearing aids (Swan et al. 2012).

A general health instrument that *may* show sensitivity to hearing aid performance is the Glasgow Benefit Inventory (GBI; Robinson et al. 1996), which has 18 questions covering the effect of a given medical intervention on physiological, personal, and social well-being. In a retrospective review, Swan et al. (2012) showed that the GBI, like the WHO-DAS II, revealed a significant benefit from provision of hearing aids, but the benefit, as for other generic instruments, was very small (+2 on a scale from −100 to +100). It is assumed that a score of 0 on the GBI is representative of no intervention, but this has yet to be validated. Overall, the generic health questionnaires are largely unresponsive to hearing impairment and its management.

10.2.4 An Analysis of Questionnaire Use

Despite the abundance of potential questionnaires that could be used for assessing the benefit of hearing aids, there are very few that are used broadly. The APHAB, COSI, GHABP, and IOI-HA have all been used many times, both in research and as part of clinical protocols (e.g., Perez and Edmonds 2012). In addition, the HHIE has been used for numerous hearing aid validation studies. Why are so few used when there so many questionnaires?

There appear to be two important factors: (1) the need to limit the clinical/experimental time required for administration and (2) the (unrealistic) hope for a self-report panacea, applicable to all types of patients, types and degrees of hearing loss, and types of hearing aids. Many of the questionnaires are lengthy: 12 of those listed in Table 10.1 have more than 50 questions; the median length is 24 questions. To reduce time, numerous shortened questionnaires have been developed, such as the SHAPIE, APHAB, and GHABP questionnaires. At the other extreme, the IOI-HA was developed with time reduction in mind (7 questions). Just 4 others have fewer than 10 questions. But do these shortened questionnaires have enough validity and power to demonstrate benefits or deficits produced by hearing aids, especially when used to evaluate effectiveness for individual patients? That needs be demonstrated; the construct, content, and criterion validations that have been done for the original questionnaires have not always been redone for the shortened versions. The psychometric validation of hearing questionnaires remains an issue.

The hope of creating a panacea questionnaire for evaluating the benefit of hearing aids is comparable to the (vain) hope of creating a panacea hearing aid (cf. Naylor 2005): how can a single questionnaire capture the myriad ways in which a hearing aid(s) might be helping an individual, especially when many new features are being added? (See Launer, Zakis, and Moore, Chap. 4 and Mecklenburger and Groth, Chap. 5.) There are two incomplete approaches to alleviating this problem: (1) have an exhaustive (and therefore lengthy) number of situations, such as in the HAPI, or domains, such as in the questionnaire of Meister et al. (2005); and (2) attempt to develop questions that are wide enough to cover most situations encountered by most individuals, yet not so wide that they are too general and have little experimental or clinical discriminatory power.

Even if there is not a single questionnaire that can be a panacea, there does seem to be a very limited number of questions/situations that appear in many questionnaires. A recent analysis of questionnaires based on the international classification of functioning (ICF; WHO 2001a) by Granberg et al. (2014) demonstrates this clearly. They analyzed hearing questionnaires that have been standardized based on particular criteria, including many of those listed in Table 10.1. Each question in each questionnaire was analyzed for its "latent" meaning and association with the most specific relevant ICF category. Despite such specificity and the 100+ ICF categories associated with hearing, the large majority of questions occupy very few general domains: activities and participation—listening (35 % of all questions), body functions—hearing functions (29 %), body functions—auditory perception (28 %), and personal factors (26 %). Granberg et al. admit that there is an overlap between the ICF categories "hearing functions" and "auditory perception." Although this provides further evidence that general health domains—like general health self-report instruments—may not be suited for hearing aid validation, it also calls into question the need for more questionnaires, especially any generic hearing inquiry, as the collated questions all seem to ask the same few things.

10.3 Benefits

10.3.1 Calculating Self-Report Benefit

One way to validate hearing aid benefit is to compare listening difficulty without and with hearing aids. This is the common approach with objective measures, where benefit is defined as the difference between aided and unaided performance, be it in percent correct, signal-to-noise ratio, or sound localization accuracy. For subjective measures, the benefit is often reported as an absolute value (e.g., the IOI-HA). The response may be given along a scale with semantic anchors, usually from "no help at all" (or some variation thereof) to "completely helpful." The primary issue with benefit assessed in this way is understanding the magnitude of the benefit associated with a single response; for example, it is not clear what distinguishes "some help," "quite helpful," and "great help"—the three intermediate responses in the GHABP questionnaire. Nor is it certain that the terms mean the same thing to every individual.

Other questionnaires use differences between pre- and postfitting to measure hearing aid benefit (e.g., the PHAB and APHAB). The measures can be either perceived ability or disability ratings. For the GHABP, Gatehouse (1999) measured both reported benefit and what he called "derived benefit": the difference between residual disability—that is, the difficulty remaining after being fitted—and initial disability. Derived benefit is a common measure across many questionnaires, including Tannahill's (1979) use of the HHS, the APHAB, and the SAC-Hx questionnaires. Derived benefit suffers from the same interpretational issue as reported benefit, namely, it is not clear what a given change or benefit value actually represents. However, it is assumed that a positive shift, that is, a statistically significant, nonzero derived benefit, indicates some form of valid benefit. An interesting difference between derived benefit and reported benefit was reported by Humes et al. (2001). Their factor analysis of the results of a test battery incorporating numerous hearing aid verification and validation measures for 173 first-time hearing aid users showed that whereas absolute benefit was related only to the factor "subjective benefit," derived benefit was also related to the factor "handicap reduction." Thus, the difference measure captures more than the absolute measure, possibly owing to the correlation between the unaided disability component of derived benefit and handicap (Whitmer et al. 2014).

Another way of calculating benefit is to consider it in a similar way to economic decision making. In an examination of the uptake of one versus two hearing aids, Cox et al. (2011) used the notion of net benefit to explain the "rational noncompliance" (Stewart and DeMarco 2010) of those who opted for one hearing aid when they would have received objective benefit (in terms of hearing performance) from wearing two. Net benefit is not the difference between performance pre- and postfitting but the difference between reported or derived benefits and reported burdens (Stewart and DeMarco 2005). Burden is the entire cost—stigma and inconvenience as well as financial—of the hearing aid(s), as opposed to the handicaps or restrictions of the hearing loss. For example, two hearing aids are more awkward to handle than one and require more maintenance and more batteries. Net benefit is usually discussed in terms of monetary cost effectiveness, but this is only one of many possible burdens. The question is then how can hearing aid burden best be evaluated: Can an attitudinal questionnaire such as the NRHA be used to estimate burden, or do the perceived burdens need to be extracted directly from the patient and quantified, and depending on the provision, does willingness-to-pay need to be considered?

10.3.2 An Example of Reported, Derived and Net Benefit

To understand better the potential uses of derived benefit and its relationship to absolute benefit, a large sample of GHABP responses is used. These data were previously used to report norms based on better-ear pure-tone average hearing loss (Whitmer et al. 2014). In total, 1,574 adults were given the GHABP as part of a general screening visit to the MRC Institute of Hearing Research in Glasgow. Of those, 997 wore either one or two hearing aids. For each of the four speech-related situations of the GHABP, participants responded as to whether the situation occurred for them and rated their difficulty hearing without their hearing aids (initial disability), the help their hearing aids provided (reported benefit), and their difficulty hearing with their hearing aids (residual disability). There are five response choices for each question, which are quantified on an interval scale (1 = "No difficulty," 2 = "Only slight difficulty," 3 = "Moderate difficulty," 4 = "Great difficulty," 5 = "Cannot manage at all"). The derived benefit, as defined by Gatehouse (1999), is the initial disability minus the residual disability and so can range from −4 (maximum possible deficit) to +4 (maximum possible benefit). A positive derived benefit indicates that the hearing aid has alleviated some difficulty. Derived benefit is limited by the amount of initial disability reported and implicitly assumes uniform steps along the subjective scale (e.g., a shift from "great" to "only slight" difficulty is assumed to be equivalent to a shift from "moderate" to "no" difficulty). Despite these limitations, derived benefit can demonstrate differences not apparent in reported absolute benefit.

Figure 10.1 illustrates the results. The left column shows, for each situation, the frequency of responses to each possible answer (1–5) for reported absolute benefit on the GHABP for unilaterally and bilaterally fitted participants (white and black bars, respectively). The distributions were similar for the unilaterally and bilaterally fitted participants, as was the mean response for each situation (shown as open and closed triangles, respectively; triangles overlap in the middle panels). The right column shows the corresponding data for derived benefit. These data reveal a difference in distribution as well as mean response: there is more positive skew for the bilaterally fitted than for the unilaterally fitted participants. This skew indicates an overall benefit of bilateral fitting. It decreases, however, with increasingly complex situations, becoming negligible for the noisy and group situations. That is, the benefit from hearing aids reduces with increasing complexity of the environment. This has been shown repeatedly with questionnaires involving different scenarios (e.g., Walden et al. 1984).

Of the 997 aided participants, 813 also completed the Hearing Handicap Questionnaire (HHQ; Gatehouse and Noble 2004), which asks them to rate annoyance and restrictions while aided, on a five-point scale. The mean aided handicap from the HHQ was used as a surrogate for treatment burden; net benefit was calculated as reported benefit minus average handicap. The distribution of responses is shown in Fig. 10.2. Mean unilateral and bilateral net benefit decreased for the noise and group situations, with bilateral net benefit always

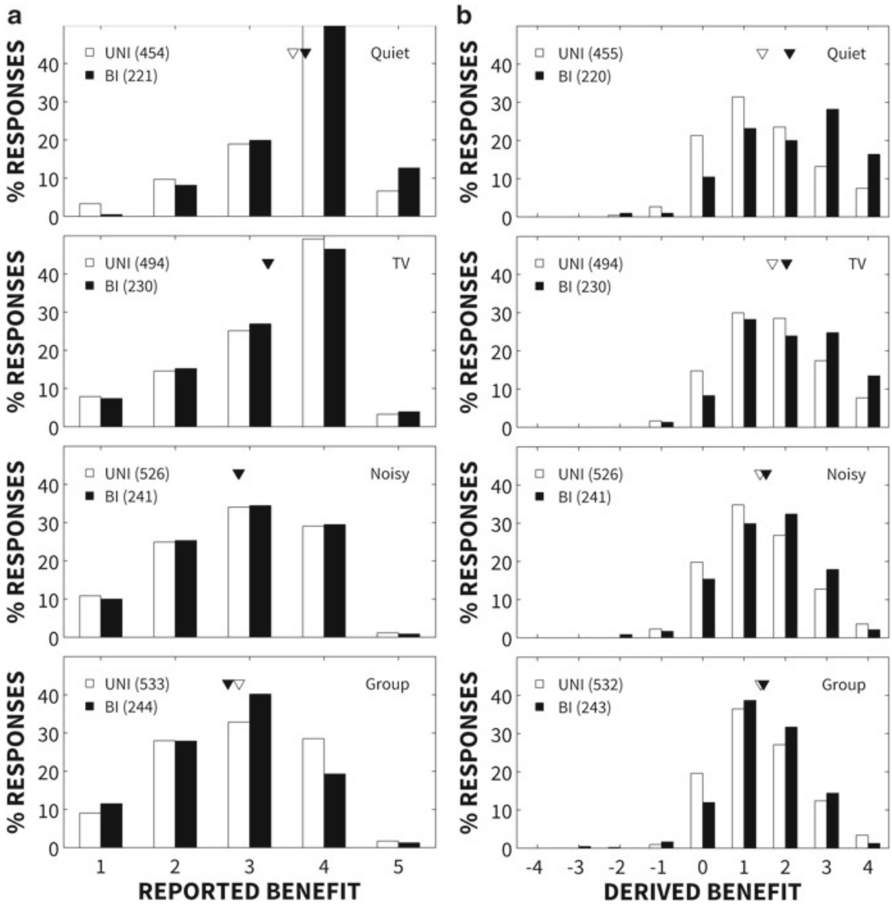

Fig. 10.1 (**a**) Reported benefit response distributions for unilaterally (*open bars*) and bilaterally (*closed bars*) fitted respondents to the GHABP in four scenarios. (**b**) Derived benefit distributions for the same respondents, calculated from the difference between initial and residual disability. Mean responses in each situation are indicated with *open* and *closed triangles*

lower than unilateral net benefit (cf. Cox et al. 2011). The distributions are more markedly different than those for reported or derived benefit: whereas the peak in the distribution of responses for unilateral users is at zero for all situations, the distributions for bilateral users are negatively skewed, with peaks below zero for all situations except speech in quiet, indicating deficits, not benefits. Although this analysis used annoyance and restrictions while aided to estimate burden and the sample comprised participants who had already made their fitting decisions, it does bear out the supposition of Cox et al. (2011) that unilateral fittings may be rational based on the perceived benefits and burdens of one versus two hearing aids.

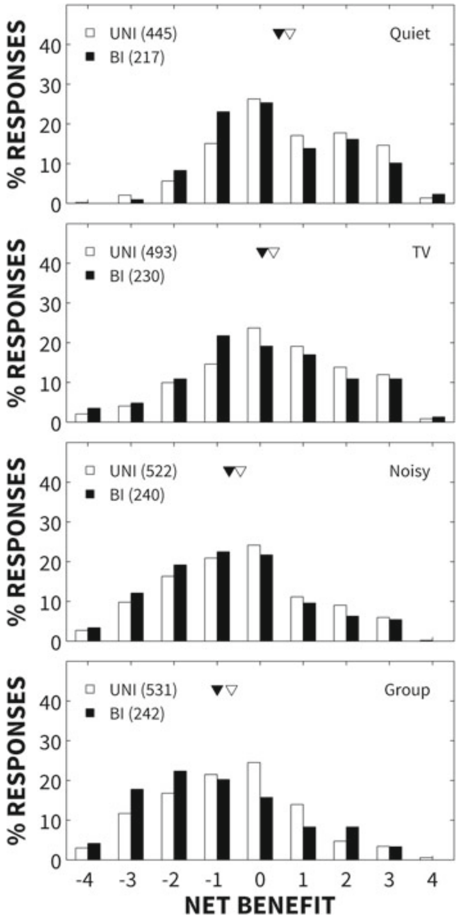

Fig. 10.2 Net benefit response distributions, calculated from the difference between aided benefit from the GHABP and aided handicap from the HHQ, for unilaterally (*open bars*) and bilaterally (*closed bars*) fitted respondents to the GHABP in four scenarios. Mean responses in each situation are indicated with *open* and *closed triangles*

10.3.3 Satisfaction and Benefit

In attempts to account for individual needs and also to emphasize the patient's concerns, satisfaction has become a vital aspect of validation. That is, satisfaction with benefit matters more than the magnitude of benefit, at both the patient and system levels. But is satisfaction with benefit really different from benefit itself? Or is the degree of satisfaction with hearing aid(s) equivalent to the amount of benefit received from the hearing aid(s)?

To investigate these issues, the correlation between (reported) benefit and satisfaction responses in the same GHABP data obtained from 997 hearing aid respondents was examined. The results are shown in Fig. 10.3. There is a strong correlation between benefit and satisfaction responses (r=0.69, 0.72, 0.75, and 0.77 for quiet,

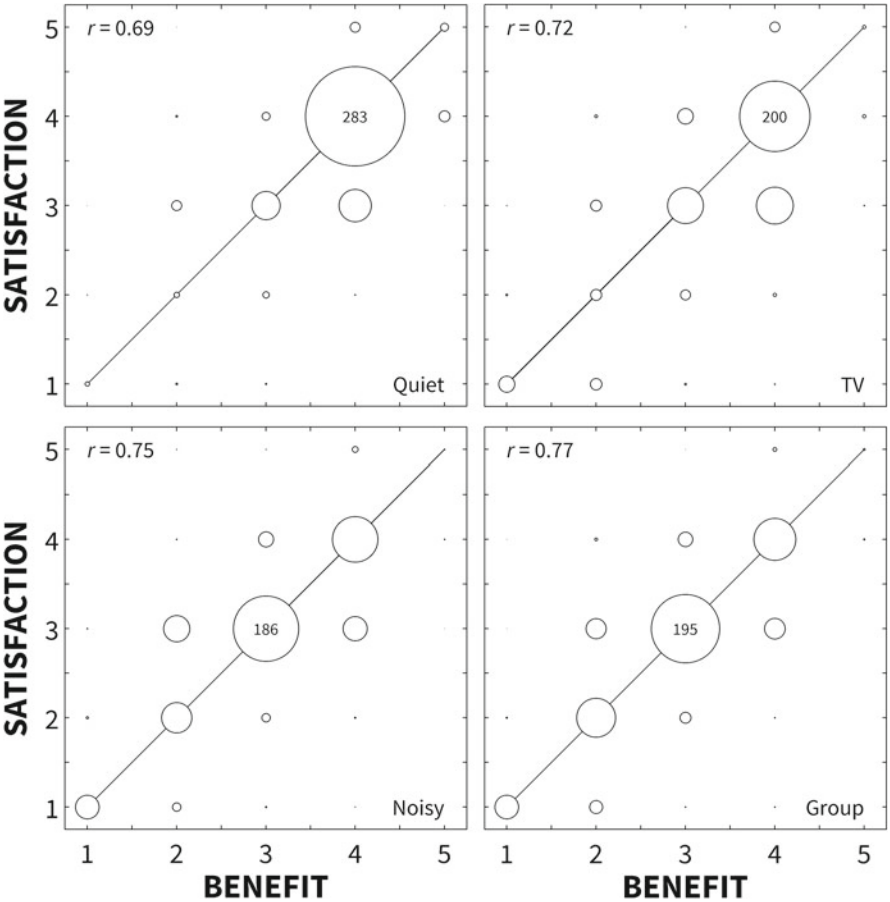

Fig. 10.3 Bubble plot of individual hearing aid satisfaction versus hearing aid (reported) benefit responses in the GHABP for all four situations (*separate panels*). Numbers indicate the number falling in the largest bubble in each panel

TV, noise, and group conversation, respectively). There is also very little skew in the responses, with 61 % of all satisfaction ratings being identical—in terms of the 1–5 scale—to (reported) benefit ratings (note that the semantic anchors are different; for use, they go from "no use at all" to "hearing is perfect with aid," whereas for satisfaction they go from "not satisfied at all" to "delighted with aid"). Despite these differences, the analysis shows a clear relationship between the two measures.

There is, however, a vital distinction between benefit and satisfaction, as stated by Demorest (1984) and illustrated in Fig. 10.4. When expectations of hearing aid performance are high, benefit, both subjectively and objectively measured, can occur together with little satisfaction. When expectations are low, there can be satisfaction yet very little benefit. Satisfaction and benefit *together* can be achieved only when

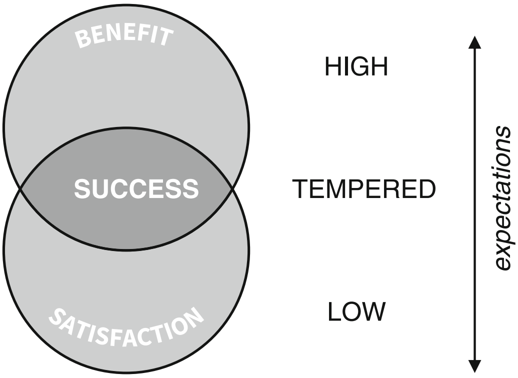

Fig. 10.4 Schematic of how the relationship between benefit and satisfaction is altered by expectations. Self-reports of satisfaction with low benefit occur when expectations (for hearing aid benefit) are low. Self-reports of benefit with low satisfaction occur when expectations are high

expectations are tempered (moderate). The distinction between satisfaction and benefit is highlighted by the difference in longitudinal effects for the two. Many studies show that satisfaction, but not benefit, decreases over time (e.g., Bentler et al. 1993). Satisfaction scores may be high at the outset and decrease over time (Tobin et al. 1997). Benefit scores, conversely, may improve over time (Cox and Alexander 1992). Tempering expectations can reduce the negative shift in satisfaction (Taylor 1993).

The distinction between satisfaction and benefit may explain how a hearing aid giving the most objective benefit is not necessarily preferred. Schum (1999) found that expectations of benefit consistently exceeded reported benefit. He also found that, across 42 new hearing aid users, there was no consistent relationship between benefits and prefitting expectations or needs. Hence, expectations do not predict benefits. Rather, they predict if benefits will be associated with satisfaction, leading to a successful—validated—hearing aid fitting. Note that this is different from the application of cognitive dissonance in consumerism, where the expectation–disconfirmation model posits that if expectations are very different from outcomes, there is a problem regardless of whether expectations fall above or below outcomes. Returning to the GHABP data discussed earlier in this section (Fig. 10.3), do the majority of respondents have tempered expectations? This was unfortunately not measured. All that is known is that they were at least occasional users of hearing aids for a period of time ranging from months to decades prior to testing; otherwise they would not have been given the GHABP questionnaire. It is plausible then that at least some of their expectations were met.

10.3.4 Validation by Use

A priori, it would seem obvious that the amount of time that someone uses their hearing aid would provide a good measure of benefit. The argument is that if they are using aid(s), there is some benefit, but if they are not (especially if they tried aids but then stopped), then there is little or no benefit. Hearing aid usage can now be measured via the data-logging capability found in many modern aids [the technology has moved far from bodyworn clocks (e.g., Haggard et al. 1981) or the weighing of batteries]. Nevertheless, there is a basic issue with validation by use: should it be weighted by the benefit or satisfaction derived during that use? That is, should a hearing aid user who wears his or her devices for 2 h a week but finds those 2 h to be highly beneficial be assessed as more or less successful than a hearing aid user who wears his or her devices all day but receives little benefit from the long hours of use? Weighting by benefit is appropriate, if only by analogy with many other healthcare interventions (e.g., many people only wear glasses or use crutches when they need them). There are few nondrug, nonsurgical interventions that give no choice but to use them all the time—a tooth filling is an example. The distinction here, between *use* and *usefulness*, limits hearing aid use per se as a measure of benefit or validation. However, combining self-report data on the situations in which a hearing aid would be useful (e.g., via the COSI) with objective data logging on the situations where the hearing aid was actually used (given that many hearing aids can classify acoustic situations; see Launer, Zakis, and Moore, Chap. 4 and Zakis, Chap. 8) as well as with what it was doing could give a new means for determining the benefit that hearing aids give an individual in the environments that matter to the individual. Given advances in hearing aid and fitting-software sophistication, this multiple-input method of measuring use for validation could be streamlined into a short process performed within the time frame of a normal consultation.

10.4 Objective Validation

Most objective measures of hearing aid performance are considered verification, ensuring that the hearing aid acoustic output is as expected. To ensure that the hearing aid meets the user's needs—that it provides benefit—requires an objective validation that probes the everyday listening of the user. The approach has been to test the most important situation for the user, which in terms of difficulty and complaints, has been understanding speech in the presence of competing noise (cf. McCormack and Fortnum 2013). While numerous speech-in-noise tests have been developed, the general—and longstanding (e.g., Carhart 1965)—problem is relating the results of these tests to everyday listening and self-reported benefit. The advantage of attempting an objective validation is that the results have a clear quantifiable interpretation, as the units of benefit are either a percentage change in speech score or a decibel change in the signal-to-noise ratio (SNR) for a fixed level of speech understanding. A subjective validation, such as a questionnaire, can lack this quantifiable interpretation, as the measurement is often inherently ordinal. The primary issue for objective validation is to mimic real-world listening tasks. This section examines recent methods in the objective validation of speech-in-noise performance that potentially bridge the gap between quantifiable listening and everyday listening.

10.4.1 Recent Advances in Speech-in-Noise Testing

Although speech-in-noise tests vary (e.g., in whether the procedure is fixed or adaptive, how limited the speech material is, or the amount of context), most tests involve the audio presentation of a sentence (the signal, S) simultaneously with noise (N, either broadband noise with a spectrum matching that of the target speech or multitalker babble). The content of the signal is repeated back by the listener after each presentation. In adaptive methods, either the S or the N is varied until a criterion percent correct is determined, generally 50 % with results recorded as an SNR. Everyday listening, conversely, is often accompanied by visual information such as lip movements, often occurs in reverberant spaces, and is continuous.

Audiovisual speech tests have been available for more than 40 years (e.g., Ewertsen and Nielsen 1971; Boothroyd et al. 1988). Two audiovisual tests with some face validity are the audiovisual version of the Connected Speech Test (CST; Cox et al. 1989) and the Sentence Gist Recognition test (SGR; Tye-Murray et al. 1996). The CST maintains context by having sentence lists culled from a passage on a (single-word) topic that is presented before testing. Wu and Bentler (2012) used the audiovisual version of the CST to show that when visual information was present, hearing-impaired individuals performed almost perfectly both with and without a directional microphone (cf. Sumby and Pollack 1954). Hence, no benefit of directional microphones could be demonstrated. The SGR test uses a video clip to introduce a scene, followed by audiovisual presentation of a sentence related to the scene; the listener is asked to choose an image that most closely represents the gist of the sentence, as opposed to repeating back the sentence. If incorrect, the listener then chooses from five "repair strategies": having the sentence repeated, simplified, rephrased, or elaborated or a key word repeated. Scores are based on the average number of "repair strategies" required to accurately identify the gist of the sentences. Despite the SGR test having relatively high face validity, it remains unused for the appraisal of hearing aids, even though the audiovisual equipment involved (e.g., touch screens) is now commonplace.

There are also numerous objective tests that measure everyday listening by the use of reverberation. Most are modifications to existing tests using isolated sentences, but the target speech and distractors are presented at different virtual locations via headphones (e.g., the R-HINT-E; Trainor et al. 2004). The different virtual locations are achieved by convolving the signals with nonindividualized (manikin) impulse responses. Unfortunately, headphone presentation makes these tests unsuitable for validating hearing aids (e.g., the limited space of circumaural headphones can lead to acoustic feedback). Tests using isolated sentences presented via loudspeakers in various arrangements (e.g., with noises presented from behind and the signal in front of the listener) are commonly used to verify the directional capabilities of the device(s). One test that was designed to simulate real-world acoustics is R-SPACE (Compton-Conley et al. 2004), which uses a loudspeaker array to simulate reflections. The R-SPACE test uses isolated sentences presented at a fixed level. The test shows that the benefit of directional microphones in reverberation is less than anticipated from results in nonreverberant conditions.

For an objective test to assess deficits and benefits in everyday communication, it should reflect the continuous nature of conversation. Although there are tests that do not require repetition of isolated sentences (e.g., the SGR test described earlier in this section), very few emulate the continuous nature of conversation. The Glasgow Monitoring of Uninterrupted Speech Test (GMUST; MacPherson and Akeroyd 2013, 2014) uses 9-min extracts of continuous speech (i.e., a short story) and asks listeners to monitor a scrolling transcript for any mismatches between the text and the audio. The SNR is adjusted based on the proportion of mismatches detected at 20-s intervals; the final score is the average of the SNRs at all but the first reversal. The GMUST demands ongoing attention but does not require shadowing (cf. Cherry 1953) or waiting until the end to test comprehension (cf. Tye-Murray et al. 2008). One of the goals of the GMUST is to address the issues revealed in the original data for the SSQ: the most difficult ranked situations involve *following* conversations (Gatehouse and Noble 2004). From the original data set (MacPherson and Akeroyd 2013), the scores for 43 adults with varying hearing abilities were found to be correlated with scores on a test of their ability to recall the content of the continuous speech material ($r = -0.42$; $p = 0.006$), reading rate ($r = -0.46$; $p = 0.002$), and SSQ responses averaged across the six conversation-following questions ($r = -0.35$; $p = 0.02$) but not with scores on a standard adaptive isolated sentence test (the Adaptive Sentence List test; MacLeod and Summerfield 1990) ($r = 0.27$; $p > 0.05$). Hence, the GMUST has potential applications in probing difficulties encountered with ongoing speech understanding that are not measured by more conventional speech perception tests. However, the GMUST has not yet been applied to assessment of the benefits of hearing aids.

The Simulated Cocktail Party (SCP) test developed by Hafter et al. (2012) is also aimed at simulating everyday listening but is primarily a test of selective attention. The SCP test presents competing talkers from different angles, asking the listener either to attend to one talker or to monitor multiple talkers (e.g., eavesdropping on another talker). Two alternative questions regarding ongoing speech material (short stories) appear visually without interrupting the story, using synonyms to probe semantic as opposed to phonetic understanding. Initial results showed that for two talkers, one primary and one secondary, answers to questions regarding the secondary talker decreased as the spatial separation between the two talkers increased. For three talkers presented symmetrically about the midline, increased separation of the sources affected scores only when all questions were about the primary (on-axis) talker's story. Like the GMUST, the SPC test has yet to be applied to assessment of the benefits from hearing aids.

An as yet unpublished "cafeteria study" test by Cohen and colleagues at Walter Reed National Military Medical Center goes beyond modeling everyday listening by having four listeners all wearing hearing aids engage in scripted conversation in an actual cafeteria; scripted questions are displayed and responses are entered by means of handheld touchscreens. Responses, reaction times and, with additional equipment, head movements presumably used to optimize spatial hearing can all be measured. Having listeners actively participate in the test in this call-response fashion may be important for assessing everyday benefit, as Erber (1992) showed that when participants initiated the sentence to be tested (in an otherwise unconnected, interrupted speech test) with a question—even just "Why?"—performance improved. Hence, tests using isolated sentences, which have been overwhelmingly prevalent in hearing aid studies, fail to reflect the multimodal, acoustic, and semantic properties of everyday conversation and do not reflect any aspect of everyday communication save for the use of words.

10.4.2 Measuring Expectation

Objective validation can also be performed via self-report: a subjective evaluation using objective methods. Based on the findings of Speaks et al. (1972) that subjective ratings of speech intelligibility were well correlated with objective scores, Cox and McDaniel (1989) developed the Speech Intelligibility Rating (SIR) test: passages composed of concatenated CST sentences are scored on a scale from 0 to 100, representing the percentage of words understood. A revised version of the SIR test (Speaks et al. 1994) balanced the passages for intelligibility. The SIR test has been shown to be sensitive to different hearing aid gain schemes (McDaniel and Cox 1992) and to different directional microphone patterns (Saunders and Kates 1997). Cox et al. (1991b) compared normal-hearing and hearing-impaired listeners using the SIR test with audio and audiovisual stimuli. For the latter, while objective–subjective correlations for audio and audiovisual speech understanding were very high ($r=0.90$ and 0.92, respectively), the linear regression line did not cross the origin. Subjective scores of 0 for audio and audiovisual speech corresponded to objective scores of approximately 12 and 15 rationalized arcsine units (transformed percentages), respectively. That is, subjective scores were, in general, lower than objective scores.

Discrepancies between measured and self-reported speech-in-noise performance can also be used to evaluate perceived disability or limitations and to validate hearing aid use. To this end, Saunders and Cienkowski (2002) developed the Perceptual Performance Test (PPT). In the PPT, an adaptive speech test using isolated sentences (the Hearing In Noise Test; Nilsson et al. 1994) is used to estimate the SNR at the speech reception threshold (SRT, the SNR required for 50% speech understanding). In a separate procedure, the SNR of sentences from the same corpus is then adjusted to when "the subject believes he or she can just understand everything that is being said" (p. 41). The difference between this subjectively adjusted SNR threshold and the SRT is the perceptual-performance discrepancy. Similar to how Demorest explained hearing aid success as being related to well-tempered expectations (1984), Saunders and Forsline (2006) were able to partially explain variation in hearing aid satisfaction as being inversely proportional to this perceptual-performance discrepancy.

A wholly subjective difference measure is the Acceptable Noise Level (ANL) (Nabelek et al. 1991). For the ANL test, a continuous passage of connected prose is presented and adjusted to the listener's most comfortable loudness level (MCL): first too loud, then too soft, then just right. Noise is added and set to a background noise level (BNL) in similar fashion: first too loud, so the passage is not understandable, then soft enough for the passage to be completely understood, then to the maximum level that the listener would "put up with" in a given situation. The ANL is the difference between the MCL and the BNL (both in decibels). As pointed out by Mueller et al. (2006), the ANL is not a level per se but an SNR, making it well suited to test hearing aid technology. The ANL test has been shown to be sensitive to changes in directional technology (Freyaldhoven et al. 2005). But unlike standard speech tests, the ANL has also shown benefits of digital noise reduction (Mueller et al. 2006) and may be a viable predictor of hearing aid use, as lower ANLs were associated with those using aids consistently (Nabelek et al. 2006). While its use for validating hearing aid features continues, its predictive power has recently come into question (e.g., Ho et al. 2013) as has its reliability (Olsen and Brännström 2014). The reliability of the ANL test may be affected by subtle modifications to the procedure (cf. reviewer comments at end of Olsen and Brännström 2014). Apart from the use of continuous discourse, there is little in the ANL test that is everyday; perhaps its predictive power or reliability would increase across sites if visual cues and/or reverberation also were presented.

10.4.3 Just-Noticeable Benefits

Several tests for assessing hearing aid benefit, both subjective and objective—see reviews by Bentler and Kramer (2000) and Saunders et al. (2009)—allow scores to be compared with a critical difference. Although this allows a practitioner to determine whether an individual's score has been improved significantly by the provision of hearing aids based on normative data, it does not indicate another important aspect of a measured difference: Is the benefit noticeable to the user of the hearing aids?

One of the most common objective validation measures is the change in SNR (unaided versus aided) at which the hearing aid user is just able to repeat all (or some fixed percentage) of the utterances (or key words therein). McShefferty et al. (2015) measured the just-noticeable difference (JND) in SNR. A sentence in noise was presented in two successive intervals. The SNR differed across the two intervals, and listeners were asked to indicate which one was clearer. An SNR improvement of about 3 dB was necessary for the higher SNR to be identified 79 % of the time. Hence, in the clinic, any adjustment to a hearing aid would need to produce a change in SNR greater than 3 dB for the client to reliably notice that change based on presentation of a single sentence. To put this into perspective, the noise reduction benefit of directional microphones is often about 3 dB (Dittberner and Bentler 2003) or even less when in realistic environments such as those simulated in the R-SPACE test (Compton-Conley et al. 2004). This does not mean that improvements less than 3 dB are unimportant, as such benefits might be appreciated across days, weeks, or months. The SNR JND provides perspective on the chasm between the subjective and objective validations described above. The hearing aid(s) may provide measurable objective speech-in-noise benefits, but if those benefits fall below the JND, there may not be a corresponding increase in self-reported benefit.

10.5 Summary

This chapter has described the myriad measurement instruments used for subjectively validating hearing aids. The only sign of progression in questionnaires in recent times has been in specialization; the domains have remained relatively fixed since the 1980s. What is clear from this review is that there is scant need for further development. It is also of note that nearly all of the questionnaires discussed have been validated to some extent. The goalposts, however, are still being moved: even the hallmark of consortium development, the IOI-HA, does not meet the most recent standards of health status measurement (i.e., the COSMIN checklist; Mokkink et al. 2010). It is thus likely that further questionnaires are inevitable, despite more than a thousand hearing aid-related questions being currently available.

The middle of the chapter has been devoted to examining the primary validation measure, benefit. What is clearly needed is an approach that captures expectations and burdens; self-reported benefits are unlikely to reflect success without knowing the expectations of the user and the perceived burdens of their hearing aid(s). Based on older concepts of expectation-tempered success (Demorest 1984) and new methods of measuring net benefit (Stewart and DeMarco 2005), a potential way forward in validating hearing aids for everyday listening is possible. To determine their "practical significance" (Carhart 1965), a subjective validation of hearing aids should include (1) the expectations of the user (e.g., the ECHO), (2) the perceived burdens by way of negative associations (e.g., the NHRA), and (3) the patient-specific benefits achieved. Combining preexisting instruments into a new approach, the "social efficiency" of the hearing aid(s) could be adequately validated.

The final section of the chapter has reviewed speech understanding measures from the perspective of everyday listening, focusing on speech-in-noise testing as (1) it is of primary importance to hearing aid users and (2) speech-in-quiet testing has limited validation potential (Carhart 1965). For objective tests to validate the benefit provided by hearing aids, the tests need to mimic everyday listening conditions. Several tests do approach ecological validity, as they include visual cues, reverberation, or continuous speech with context. The conjunction "or" is crucial: despite decades of speech-in-noise test development, only one aspect of test realism has usually been considered. Unfortunately, the provision of visual cues and context can lead to ceiling effects that prevent benefit from being measured. Furthermore, recent evidence suggests that some objective benefits may be too small to be reliably noticed, despite being statistically significant. Hence, it is important to ascertain the expectations and burdens in conditions specific to the patient in validating a "socially efficient" hearing aid.

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Chapter 11
Future Directions for Hearing Aid Development

Gerald R. Popelka and Brian C.J. Moore

Abstract Hearing aids will continue to be acoustic, customizable, wearable, battery-operated, and regulated medical devices. Future technology and research will improve how these requirements are met and add entirely new functions. Microphones, loudspeakers, digital signal processors, and batteries will continue to shrink in size to enhance existing functionality and allow new functionality with new forms of signal processing to optimize speech understanding, enhance spatial hearing, allow more accurate sound environment detection and classification to control hearing aid settings, implement self-calibration, and expand wireless connectivity to other devices and sensors. There also is potential to provide new signals for tinnitus treatment and delivery of pharmaceuticals to enhance cochlear hair cell and neural regeneration. Increased knowledge and understanding of the impaired auditory system and effective technology development will lead to greater benefit of hearing aids in the future.

Keywords Batteries • Cognitive control • Epidemiology • Fitting methods • Hearing aids • Hearing impairment • Microphones • Music perception • Real-ear measurements • Receivers • Regeneration • Self-calibration • Signal processing • Spatial perception • Speech Perception • Wireless connectivity

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11.1 Introduction

It is likely that a large proportion of future hearing aids will continue to be regulated medical devices that will be acoustic, wearable, battery operated, and intended to be worn continuously all day. Future devices must meet the existing basic requirements that include being comfortable to wear, being cosmetically acceptable, having a battery life of at least one full day, and being customizable to produce frequency- and level-dependent gains that are appropriate for the individual hearing-impaired person. They must be easily reprogrammable to compensate for changes in hearing function with aging or other factors. Future hearing aid technology and hearing aid-related research have the potential to improve how these requirements are met and to add entirely new functions.

11.2 Microphone Size and Technology

As described in Chap. 3 by Killion, Van Halteren, Stenfelt, and Warren, hearing aid microphones continue to shrink in size without sacrificing any of their already remarkable acoustic capabilities, including wide bandwidth (20–20,000 Hz), high maximum input level without overload (115 dB SPL), and low inherent noise floor (typically 25–30 dBA). Current microphones are robust and are available in extremely small packages, especially in the case of micro-electrical-mechanical systems (MEMS) microphones. Further size reductions will allow not only the possibility of producing smaller, more comfortable, and less visible hearing aids but also the ability to add additional multiple well-matched microphones on the same small ear-level devices. This opens up possibilities for having highly directional characteristics, which may be useful in noisy situations for selecting a "target" sound (e.g., a talker of interest) while rejecting or attenuating competing sounds. Biologically inspired highly directional microphones designed using silicon microfabrication are also on the horizon (Miles and Hoy 2006).

11.3 Receivers

Hearing aid loudspeakers, called receivers, also continue to shrink in size and their acoustic characteristics continue to be improved; see Chap. 3. The demands on receivers are complex. Their output requirements are related to the individual's hearing status and to the receiver location with respect to the tympanic membrane. The sound reaching the tympanic membrane is influenced substantially by the physical dimensions of the external ear canal, by where in the canal the receiver is located, and by the size and configuration of the venting to the external environment (see Moore and Popelka, Chap. 1 and Munro and Mueller, Chap. 9). The outputs of the receivers of future hearing aids will continue to be greatly affected by these factors and it will continue to be necessary to specify the real-ear output, as discussed in Chap. 9.

Most individuals with age-related high-frequency sensorineural hearing loss require no amplification of low-frequency sound but do require amplification of sound at higher frequencies. A popular type of hearing aid for such people is a behind-the-ear (BTE) device with a receiver in the ear canal and an open fitting, that is, a nonoccluding dome or earmold. With this configuration, the low-frequency sounds are heard unamplified via leakage of sound through the open fitting, while the medium- and high-frequency sounds are amplified and are dominated by the output of the receiver. Such designs are popular partly because they are physically comfortable for all-day wear, they can be "instant fit" (a custom earmold is not required), and because low-frequency sounds are completely undistorted and natural sounding. There are, however, some problems with this approach. First, the amplified high-frequency sound is delayed relative to the low-frequency sound through the open fitting, leading to an asynchrony across frequency that may be perceptible as a "smearing" of transient sounds (Stone et al. 2008). Second, for medium frequencies, the interaction of amplified and unamplified sounds at comparable levels may lead to disturbing spectral ripples (comb-filtering effects); see Stone et al. (2008) and Zakis, Chap. 8. Third, for high-input sound levels, the gain of the hearing aid is reduced, and the sound reaching the tympanic membrane is strongly influenced by the sound passing through the open fitting. In this case, the benefits of any directional microphone system or beamformer (see Launer, Zakis, and Moore, Chap. 4) may be reduced or lost altogether.

An alternative approach is to seal the ear canal with an earmold or soft dome; see Chap. 1. In this case, the sound reaching the tympanic membrane is dominated by the amplified sound over a wide frequency range, including low frequencies. This approach may be required when significant gain is required at low frequencies to compensate for low-frequency hearing loss because it is difficult to achieve low-frequency gain with an open fitting (Kates 2008). A closed fitting avoids problems associated with temporal asynchrony across frequency, spectral ripples, and loss of directionality at high levels. However, there are also drawbacks with this approach. First, low-frequency sounds may be heard as less natural than with an open fitting because of the limited low-frequency response of the receiver or because the gain at low frequencies is deliberately reduced to prevent the masking of speech by intense low-frequency environmental sounds. Second, the user's own voice may sound unnaturally loud and boomy because bone-conducted sound is transmitted into the ear canal and is trapped by the sealed fitting; this is called the occlusion effect (Killion et al. 1988; Stone and Moore 2002). There are two ways of alleviating the occlusion effect. One is to use a dome or earmold that fits very deeply inside the ear canal (Killion et al. 1988). The other is to actively cancel the bone-conducted sound radiated into the ear canal using antiphase sound generated by the receiver (Mejia et al. 2008). To the knowledge of the authors, active occlusion cancellation has not yet been implemented in hearing aids, but it may become available in the near future.

A completely different approach to sound delivery is the "Earlens" system described in Chap. 8; see also Perkins et al. (2010) and Fay et al. (2013). This uses a transducer that drives the tympanic membrane directly. The transducer is placed directly on the tympanic membrane and receives both signal and power via a light source driven by a BTF device. The light is transmitted from a light source in the ear canal to a receiver mounted on a "chassis" that fits over the tympanic membrane. The device has a maximum effective output level of 90–110 dB SPL and gain before acoustic feedback of up to 40 dB in the frequency range from 0.125 to 10 kHz. The ear canal can be left completely open, so any occlusion effect is small or nonexistent. The gain before acoustic feedback is relatively large at high frequencies because the eardrum vibrates in a chaotic manner (Fay et al. 2006) and vibration of the tympanic membrane by the transducer leads to a much smaller amount of sound being radiated from the ear canal back to the hearing aid microphone than would be the case for a conventional hearing aid (Levy et al. 2013). This can avoid the problems associated with the use of digital signal processing to cancel acoustic feedback (Freed and Soli 2006; Manders et al. 2012). The device has been undergoing clinical trials in 2014–2015 and may appear on the market soon after, if the trials are successful.

11.4 Digital Signal Processors and Batteries

Since the introduction of the first full digital hearing aid (Engebretson et al. 1985), wearable digital signal processors (DSPs) have shrunk progressively in size and power requirements, characteristics that have substantial implications for the future. Reduced power requirements together with improvements in battery technology will probably contribute to increased intervals between battery replacement or recharging. This is a consideration not only for patient convenience but also to ensure that more demanding signal processing can be accommodated without increasing battery or processor sizes. It is likely that, in the future, new effective and beneficial communication enhancement signal-processing algorithms, new automated convenience features, and new fitting and adjustment capabilities all can be added to substantially improve overall hearing aid function without increasing the physical size of the digital signal processor.

Battery technology is being actively researched, driven by rapid growth of mobile devices. The likely developments will be in both the battery chemistry and internal components such as anodes (Lin et al. 2015). Future improvements may include longer battery life, very rapid recharging for rechargeable batteries, innovative packaging to optimize space within the hearing aid casings, and possibly increased voltage that may help increase dynamic range and DSP processor speed.

Battery life and processor power consumption are also important for hearing aids that are inserted deeply into the ear canal and are intended to be left in the ear for extended periods (Palmer 2009). Currently, such devices use analog signal processing that requires less power than digital signal processing. The devices also have no wireless connectivity. At present, the devices can be left in place for 3–4 months, but improvements in battery and DSP technology could lead to longer durations of use.

Current DSPs already provide multiple adjustable channels that can compensate for sensitivity loss in a frequency-selective manner and provide amplification tailored to the individual's hearing requirements. Almost every hearing aid incorporates some form of frequency-selective amplitude compression, also called automatic gain control (AGC). However, the way in which this is implemented differs markedly across manufacturers, and it remains unclear which method is best for a specific patient, if there is indeed a “best” method (Moore 2008). There is evidence that the preferred compression speed (see Chaps. 4 and 6) varies across hearing-impaired listeners (Gatehouse et al. 2006; Lunner and Sundewall-Thoren 2007), but there is at present no well-accepted method for deciding what speed will be best for an individual. Hopefully, in the future, methods for implementing multichannel AGC will be refined and improved, and better methods will be developed for tailoring the characteristics of the AGC to the needs and preferences of the individual.

A form of signal processing that has attracted considerable interest in recent years involves frequency lowering, whereby high-frequency components in the input signal are shifted to lower frequencies for which hearing function is usually better; see Chaps. 4, 6, and 8. Most major manufacturers of hearing aids now offer some form of frequency lowering, but the way in which it is implemented varies markedly across manufacturers. Most published studies evaluating the effectiveness of frequency lowering suffer from methodological problems, and at present, there is no clear evidence that frequency lowering leads to benefits for speech perception in everyday life. Also, it remains unclear how frequency lowering should be adjusted to suit the individual, what information should be used when making the adjustments, or how long it takes to adapt to the new processing. It is hoped that, in the future, frequency-lowering methods will be improved and well-designed clinical studies will be conducted to compare the effectiveness of different methods of frequency lowering and to develop better methods of fitting frequency lowering.

Many hearing aids perform a type of “scene analysis”; see Chaps. 4 and 8. For example, they may classify the current scene as speech in quiet, speech in noise, noise alone, or music. The parameters of the hearing aid may then be automatically adjusted depending on the scene. In current hearing aids, the number of identified scenes is usually limited to about four, and the classifier is pretrained by the manufacturer, using “neural networks” and a large set of prerecorded scenes. In the future, it may be possible to identify many more types of scenes—for example, speech in background music, speech in a combination of background noise and music, music in car noise (which has most of its energy at low frequencies), speech in a reverberant setting, music in a reverberant setting, classical music, pop music, jazz music—and to adjust the parameters of the hearing aid accordingly. Possibly, as mentioned in Chap. 8, the scene classifiers could automatically learn the scenes that the individual user encounters most often.

There are some problems with the use of classifiers to control hearing aid settings. First, it is often not obvious how to adjust the hearing aid for any specific scene. For example, should the compression speed be different for speech and for music and should the frequency-gain characteristic be different for speech and for music? More research is clearly needed in this area. A second problem is more fundamental. Sensory systems generally seem to have evolved to provide accurate information about the outside world. In the case of the auditory system, the goal is to determine the properties of sound sources. This requires a consistent and systematic relationship between the properties of sound sources and the signals reaching the ears. But if a hearing aid changes its characteristics each time a new scene is identified, there is no longer a consistent relationship between the properties of the sound source and the signals reaching the ears. This may make it more difficult for a hearing aid user to interpret auditory scenes, especially when they are complex. More research is needed to assess the severity of this problem and to determine whether there are overall benefits to be obtained from the use of scene classifiers.

As speech recognition systems improve, the opportunity may develop for real-time processing specifically intended to optimize perception of individual speech sounds by modifying both the temporal and spectral aspects of the sounds. Some forms of processing of this type, implemented "by hand" on "clean" speech, have been shown to be beneficial (Gordon-Salant 1986; Hazan and Simpson 2000). However, automatic processing of this type may be extremely difficult to implement, especially when background sounds are present. Also, automatic processing may involve significant time delays (Yoo et al. 2007) and this would disrupt the temporal synchrony between the auditory and visual signals from the person speaking. The audio component of speech can be delayed by up to 20 ms before there is a noticeable and interfering mismatch between what is seen on the face and the lips of the speaker and what is heard (McGrath and Summerfield 1985). Although modern DSPs are able to perform very large numbers of computations in 20 ms, there may be intrinsic limitations in automatic processing to enhance specific speech sound or features that prevent the processing delay from being reduced below 20 ms.

11.5 Self-Calibrating Hearing Aids

Variations in the geometry of the external ear canal, the position of the receiver, and the type of seal all greatly affect the sound reaching the tympanic membrane. This usually requires real-ear measures to check and adjust the actual output of the hearing aid, as described in Chap. 9. A possible way of reducing the need for such measures is via self-measuring and self-calibrating features in hearing aids. Such features were originally proposed and implemented in the first full digital hearing aid (Engebretson et al. 1985). The self-calibration required a microphone facing inward, toward the tympanic membrane. As microphones continue to shrink in size or as other approaches emerge that do not require onboard sound measurement technology (Wiggins and Bowie 2013), self-calibrating hearing aids are likely to become more common in the future. Such systems can help in achieving target frequency- and level-dependent gains at the initial fitting and can greatly speed up the initial fitting process. In addition, they could potentially compensate for day-to-day variations resulting from, for example, cerumen accumulation and removal and different ear canal positions of the receiver resulting from removing and reinserting the device. Insertion of a hearing aid could automatically trigger a self-adjustment procedure to ensure that the device provided the desired output at the tympanic membrane.

11.6 Wireless Connectivity

As discussed in Chap. 5 by Mecklenburger and Groth, wireless connectivity for hearing aids allows significant improvements in many hearing aid functions. Wireless connectivity allows the use of remote microphones and of signal processing in devices outside the hearing aid itself. Currently, the "streamer" modules described in Chap. 5 can communicate with the hearing aids worn on each ear. The microphones on a single "streamer" must be close together and do not provide the advantages for beamforming or spatial hearing benefits of the more separated microphone locations of the hearing aids worn on each ear, as discussed in Chaps. 4, 5, and 7. Future technology may allow wireless transmission of the outputs of the microphones on the two ear-level hearing aids to the "streamer" for signal processing and then transmission back to the ear-level devices. This could allow more computationally demanding signal processing than is possible at present.

In addition to the "streamer" component, the list of external devices that current hearing aids connect to wirelessly includes mobile telephones, television sets, and remote microphones. Because mobile telephones also independently connect wirelessly to a variety of other devices, such as the audio systems in cars, the number and variety of devices connected to hearing aids will increase automatically as the list of connected devices to mobile phones increases. It is already possible for a mobile telephone global positioning system to identify a specific location (e.g., a restaurant) and to select a set of parameters in the hearing aid that have previously been found to be preferred in that situation.

Currently, there is an emphasis on new technology embedded within "wearables," small electronic devices that contain substantial computing power and sensors. Examples are glasses, wrist watches or other wrist-worn devices, and even contact lenses. Wearables may function as health and fitness monitors or medical devices. They collect physiological data from tiny sensors such as pressure sensors, temperature sensors, and accelerometers, analyze the data, and provide information to the wearer, often in real time via a visual display. Future hearing aids may incorporate such sensors or be linked wirelessly to devices containing the sensors and may present the information via an auditory speech signal tailored to the wearer. This represents only a small extension to the current capability of some hearing aids to provide a synthesized voice signal to indicate what program has been selected or to warn the user of the need to change the battery.

11.7 Tinnitus Treatment

Some hearing aids have the ability to generate sounds that may be used to mask tinnitus, to draw attention away from tinnitus, or to help the tinnitus sufferer to relax, especially when used together with appropriate counseling (Aazh et al. 2008; Sweetow and Sabes 2010). Future efforts may involve the use of hearing aids to supply digitally synthesized signals intended to reduce the severity of tinnitus using principles drawn from studies of neuroplasticity (Tass and Popovych 2012; Tass et al. 2012). The clinical effectiveness of such tinnitus intervention strategies has not yet been clearly determined. Further research is needed to determine the benefits of these approaches and to evaluate the relative effectiveness of the different types of tinnitus intervention.

11.8 Cognitively Controlled Hearing Aids

Hearing aids already exist that act as "binaural beamformers," selectively picking up sounds from a particular, selectable direction (e.g., a specific talker); see Chaps. 4 and 5. A practical problem is that the hearing aids do not "know" which source the user wants to attend to at a specific time. There is evidence that brain activity and the corresponding evoked electrical responses change depending on which sound source a person is attending (Mesgarani and Chang 2012; Kidmose et al. 2014). In principle, therefore, the beamforming in hearing aids could be controlled by evoked potentials measured from the user such that the desired source/direction was automatically selected. This has sometimes been referred to as "cognitively controlled hearing aids." There are many serious problems that need to be solved before such hearing aids become practical. A major problem is that users may switch attention very rapidly between sources from different directions. The hearing aids would need to switch almost as rapidly to avoid the directional beam "pointing" at the wrong source. Currently, considerable averaging over time is needed to extract "clean" evoked potentials from sensors on the scalp or in the ear canal (Kidmose et al. 2013). It is not known whether it will be possible to derive an evoked potential indicating the desired source signal or its direction with sufficient speed to satisfy the needs of the user. Research is currently ongoing to explore the feasibility of cognitively controlled hearing aids.

11.9 Using Hearing Aids to Enhance Regeneration

At present, many laboratories throughout the world are investigating a variety of approaches to regeneration or repair of sensory and related structures to restore auditory function. The approaches include use of stem cells and a variety of gene therapies (Izumikawa et al. 2005; Oshima et al. 2010; Rivolta 2013; Ronaghi et al. 2014). Although these approaches are beginning to show promise, none are expected to be successful in the near future.

Because of the wide variety of cochlear pathologies and genetic disorders, it is likely that a variety of approaches will emerge that are pathology specific. The biological interventions are usually designed to imitate the normal patterns of biological development. These are very complex and involve cell differentiation regulated by nerve growth factors and other chemicals that are released at very specific developmental periods.

Furthermore, normal auditory development at the cellular level requires auditory signals to guide development. Abnormal auditory input can result in abnormalities and changes at many levels in the auditory system (Sharma et al. 2007). Future hearing devices may be developed to enhance biological treatments for regeneration or repair. A future hearing aid system may include an acoustic hearing aid and a linked implanted component capable of eluting chemicals and even producing electrical signals. The system would be able to provide controlled acoustic, electrical, and pharmaceutical signals at the appropriate time to control the developmental process and, when complete, the device could be removed.

11.10 Concluding Remarks

Age expectancy is increasing, but hearing function continues to decrease with increasing age. Hence the need for hearing aids, and improvements in hearing aids, is greater than ever. Current hearing aids are effective in improving the audibility of sounds, but they remain of limited benefit in the situations in which they are most needed, namely in noisy and reverberant environments.

Knowledge and understanding of the impaired auditory system continue to improve, and effective technology development continues. Hopefully, this will lead to greater benefit of hearing aids in the future and to a much greater extent of hearing aid use among those with hearing loss.

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