

Hearing Aids

Introductory Comments

Hearing aids partially overcome the deficits associated with a hearing loss. For a sensorineural hearing loss, there are several deficits to be overcome. Some sounds are inaudible. Other sounds can be detected because part of their spectra is audible, but may not be correctly identified because other parts of their spectra (typically the high-frequency parts) remain inaudible. The range of levels between the weakest sound that can be heard and the most intense sound that can be tolerated is less for a person with sensorineural hearing loss than for a normal-hearing person.

To compensate for this, hearing aids have to amplify weak sounds more than they amplify intense sounds. In addition, sensorineural impairment diminishes the ability of a person to detect and analyze energy at one frequency in the presence of energy at other frequencies.

Similarly, a hearing-impaired person has decreased ability to hear a signal that rapidly follows, or is rapidly followed by, a different signal. Hearing-impaired people are also less able to separate sounds on the basis of the direction from which they arrive. This decreased resolution (frequency, temporal, and spatial) means that noise, or even other parts of the speech spectrum, will mask speech more than would be the case for a normal-hearing person.

The physiological origins of sensorineural hearing loss include loss of inner hair cell function, outer hair cell function, reduced electrical potential within the cochlea, and changes to the mechanical properties of the cochlea. The resulting auditory deficits mean that a person with a sensorineural hearing impairment needs a signal-to-noise ratio greater than normal in order to communicate effectively, even when sounds have been amplified by a hearing aid. In contrast, a conductive impairment simply attenuates sound as it passes through the middle ear, so the amplification provided by hearing aids comes close to restoring hearing to normal.

To understand how hearing aids work, the physical characteristics of signals must be understood. These characteristics include the rate at which sound fluctuates (frequency), the time taken for a repetitive fluctuation to repeat (period), the distance over which its waveform repeats (wavelength), the way sound bends around obstacles (diffraction), the strength of a sound wave (pressure and sound pressure level), the break-up of a complex sound into pure tone components at different frequencies (spectrum), or into several frequency bands (octave, one-third octave or critical bands), and the degree to which a body of air vibrates when it is exposed to vibrating sound pressure (velocity and impedance).

The amplifiers inside hearing aids can be classified as linear or nonlinear. For sounds of a given frequency, linear amplifiers amplify by the same amount regardless of the level of the signal, or what other sounds are simultaneously present. By contrast, the amplification provided by a nonlinear amplifier varies with the amplitude of the signal input to the amplifier. The degree of amplification can be represented as a graph of gain versus frequency (gain-frequency response), or as a graph of output level versus input level (I-O curve). The highest level produced by a hearing aid is known as the saturation sound pressure level (SSPL). SSPL is usually estimated by measuring the output sound pressure level for a 90 dB SPL input (OSPL90).

The sound output by a hearing aid can be measured in the ear canal of an individual patient, or in a small coupler or ear simulator that has a volume similar to that of a real ear.

Hearing aids are described according to where they are worn. In order of decreasing size these categories are: body, spectacle, behind-the-ear, in-the-ear, in-the-canal and completely-in-the-canal. For behind-the-ear hearing aids, further categorization is needed to distinguish between styles where the hearing aid receiver (the output transducer) is within the hearing aid case or within the ear canal.

Decreasing size has been a constant trend during the history of the hearing aid. This history can be divided into six eras: acoustic, carbon, vacuum, transistor, digital, and wireless. The last of these eras, which we are just entering, promises to hold advances at least as significant as in the eras that preceded it.

Hearing Aid Components

Hearing aids are best understood as a collection of functional building blocks. The manner in which a signal passes through these blocks in any particular hearing aid is indicated in a block diagram. The first block encountered by an acoustic signal is a microphone, which converts sound to electricity. Modern miniature electret microphones provide a very high sound quality, with only very minor imperfections associated with internal noise and sensitivity to vibration. Directional microphones, which have two entry ports, are more sensitive to frontal sound than to sound arriving from other directions. These enable hearing aids to improve the signal-to-noise ratio by several decibels (depending on acoustic conditions) relative to omnidirectional microphones, and hence can improve the intelligibility of speech in noise. Dual-microphone hearing aids can be switched automatically or by the user to be either directional, or omni-directional, as required in different listening situations.

The small signals produced by microphones are made more powerful by the hearing aid amplifier. All amplifiers will distort the signal, by peak clipping it, if they attempt to amplify the signal to too high a level. Excessive distortion decreases the quality and intelligibility of sounds. To avoid distortion, and to decrease the dynamic range of sound, compression amplifiers are used in most hearing aids. These amplifiers decrease their gain as the level of the signal put into them increases, in much the same way that a person will turn down a volume control when the level becomes too high.

Amplifiers can represent sound in an analog or a digital manner. The signals within analog amplifiers have waveforms that mimic the acoustic waveforms they represent. Digital systems represent signals as a string of numbers. Performing arithmetic on the string of numbers alters the size and nature of the signals these numbers represent. Fully digital circuits may be constructed so that they process sounds in ways specific to each device, or may be able to perform any arithmetic operation, in which case the type of processing they do depends on the software that is loaded into them.

Filtering a signal is a common way in which hearing aids alter sound. Filters can be used to change the relative amplitude of the low-, mid- and high-frequency components in a signal. When the filters are made with variable, controllable characteristics, they function as tone controls operated by the user or the clinician. Filters can also be used to break the signal into different frequency ranges, so that different types of amplification can be used in each range, as required by the hearing loss of the hearing-impaired person.

Receivers are miniature headphones that use electromagnetism to convert the amplified, modified electrical signals back into sound. Their frequency response is characterized by multiple peaks and troughs, which are partly caused by resonances within the receivers, and partly caused by acoustic resonances within the tubing that connects a receiver to the ear canal. Inserting an acoustic resistor, called a damper, inside the receiver or tubing will smooth these peaks and troughs. A damper absorbs energy at the frequencies corresponding to the peaks, and this improves sound quality and listening comfort.

There are several other ways to put signals into hearing aids. A telecoil senses magnetic signals and converts them to a voltage. A radio receiver senses electromagnetic waves and converts them to a voltage. A direct audio input connector enables an electrical audio signal to be plugged straight into the hearing aid.

Users operate hearing aids via electromechanical switches on the case of the hearing aid, or via a remote control. The hearing aid performs all its functions by taking electrical power from a battery. These batteries come in a range of physical sizes and capacities, depending on the power needed by each hearing aid, and the space available.

Hearing Aid Systems

Components can be combined into hearing aids in an extremely customized manner, such that individual components are selected for each patient and are located in the position that best suits each ear. At the other extreme are modular aids, including some ITC hearing aids, and all BTE hearing aids, that are prefabricated in a totally standard manner. Many hearing aids fall somewhere between these extremes.

Increasingly, the hearing aids on each side of the head communicate with each other by wireless transmission so that their amplification characteristics (directionality, noise reduction processing, compression characteristics and input source) can remain coordinated as the environment changes or as the user varies a control. In some cases, the complete audio signal is transferred from one side of the head to the other, which enables a telephone signal to be heard in both ears, and will enable super-directional microphones to be developed.

Hearing aid amplification characteristics are programmed from a computer, via a suitable wired or wireless interface, to suit the hearing capabilities of each user. Often, more than one program is put into the hearing aid so that different amplification characteristics can be selected, automatically by the hearing aid or manually by the user, in different listening conditions.

The most effective way to make speech more intelligible is to put the microphone near the lips of the person talking. This markedly decreases noise and reverberation, but requires a means of transmitting the signal from the microphone to the hearing aid wearer some distance away. Methods to do this currently include (1) magnetic induction from a loop of wire to a small telecoil inside the hearing aid, (2) radio transmission of a frequency-modulated, or digitally modulated electromagnetic wave, (3) infrared transmission of an amplitude-modulated electromagnetic wave, and (4) acoustic transmission of an amplified sound wave. Each of these systems has strengths and weaknesses compared to the others. The first three offer a very large potential increase in signal-to-noise ratio, and hence intelligibility. It can, however, be a challenge to adjust the hearing aid and

wireless system together so that both the wireless system and the hearing aid individually provide maximum benefit to the wearer without the wireless input and microphone input signals interfering with each other. Increasingly, wireless receivers are being built into hearing aids, considerably improving cosmetic appearance and convenience. A major application of these systems is to make teachers more easily understood in classrooms, but they can be used by children and adults in other situations as well.

Wireless reception is also enabling hearing aids to conveniently accept electrical signals from a range of audio devices, including televisions, MP3 players, computers and mobile phones. In many cases, this connection requires an intermediary wireless relay device, as the current consumption of the ubiquitous Bluetooth receivers and transmitters precludes them from being directly built into hearing aids. While there have been problems with mobile phones causing interference in hearing aids, this problem is decreasing due to improvements in hearing aid design and changes in the mobile phone transmission system. Use of a mobile phone via hearing aids is now often trouble-free. There is the potential for hearing aids to become the audio portal to the world, and possibly not just for hearing-impaired people.

Assistive listening devices enable hearing aid wearers to receive sounds other than just by the amplification provided in a self-contained hearing aid. Assistive listening devices include the transmitter/receiver pairs already described for remotely sensing and sending speech or music, and devices that alter sound at its source (such as a telephone amplifier). Other types of assistive listening devices enable the aid wearer to detect alerting sounds (e.g. the doorbell, a telephone ring, a smoke alarm). Some do this by transmitting the sounds wirelessly to the hearing aids; others convert sound to other sensory modalities (such as flashing lights or vibrating shakers).

The long-established distribution and fitting system for hearing aids is being somewhat challenged by over-the-counter hearing aids, and their more modern cousin, hearing aids sold over the internet, and even by disposable hearing aids.

Electroacoustic Performance and Measurement

The performance of hearing aids is most conveniently measured when the hearing aid is connected to a coupler. A coupler is a small cavity that connects the hearing aid sound outlet to a measurement microphone. Unfortunately, the standard 2-cc coupler is larger than the average adult ear canal with a hearing aid in place, so the hearing aid generates lower SPL in this coupler than in the average ear. This difference is called the real-ear-to-coupler difference (RECD) a quantity that is worth measuring in infants because they have ear canals considerably different from the average adult. A more complex measurement device, which better simulates the acoustic properties of the average adult ear canal, is called an ear simulator.

Test boxes provide a convenient way to get sound into the hearing aid in a controlled manner. These sounds can be pure tones that sweep in frequency, or can be complex, broadband sounds that, like speech, contain many frequencies simultaneously. Broadband sounds are necessary to perform meaningful measurements on many nonlinear hearing aids. Increasingly, it is necessary for the test sound to approximate the spectral and temporal properties of speech so that the various signal processing algorithms in the hearing aid alter the gain in a manner representative of actual use. The measurements most commonly performed using test sounds are curves of gain or output versus

frequency at different input levels, and curves of output versus input at different frequencies. The curve of output versus frequency when measured with a 90 dB SPL pure tone input level is usually taken to represent the highest levels that a hearing aid can create. Some other test box measurements that are less commonly performed are measures of distortion, internal noise, and response to magnetic fields. These measurements are used to check that the hearing aid is operating in accordance with its specifications.

Test box measurements are but a means to an end. That end is the performance of the hearing aid in an individual patient's ear. This performance can be directly measured using a soft, thin probe-tube inserted in the ear canal. Real-ear performance can be expressed as real-ear aided response (REAR; the level of sound in the patient's ear canal), real-ear aided gain (REAG; the level of sound in the ear canal minus the input level of sound near the patient) or as real-ear insertion gain (REIG; the level of sound in the ear canal when aided minus the level in the same place when no hearing aid is worn). Each of these measures requires the probe to be carefully located, but the requirements for probe placement are a little less critical for REIG than for REAG or REAR.

Both types of real-ear gain are different from coupler gain, partly because of the real-ear-to-coupler difference already mentioned, and partly because the input to the hearing aid microphone is affected by sound diffraction patterns around the head and ear. The changes in SPL caused by diffraction are referred to as microphone location effects. Insertion gain is further different from coupler gain because resonance effects in the unaided ear form a baseline for the insertion gain measurement. This baseline, referred to as the real-ear unaided gain, provides the link between the REAG and the REIG.

Many factors can lead to incorrect measurement of real-ear gain. These factors include incorrect positioning of the probe, squashing of the probe, blockage of the probe by cerumen, background noise, and hearing aid saturation. Fortunately, there are some simple checks one can do to verify measurement accuracy.

Feedback oscillation is a major problem in hearing aids. It happens when the amplification from the microphone to the receiver is greater than the attenuation of sound leaking from the output back to the input. Clinicians must be able to diagnose the source of excess leakage. Other problems that often have simple solutions include no sound output, weak output, distorted output, and excessive noise.

Hearing Aid Earmoulds, Earshells and Coupling Systems

An earmould or earshell is moulded to fit an individual's ear and retains the hearing aid in the ear. Pre-moulded canal fittings, available in a range of standard sizes and shapes, are an alternative way to connect the hearing aid to the ear canal. Whether custom moulded or pre-formed the ear fitting retains the sound bore, which is a sound path from the receiver to the ear canal. In many cases the fitting provides a second sound path, referred to as a vent, between the air outside the head and inside the ear canal. Where no vent exists, as in high-gain hearing aids, the fitting is said to be occluding. Where the cross-section of the ear canal remains largely unfilled for its entire length, the hearing aid is said to be an open fitting, or an open-canal fitting. The three functions of an ear fitting are thus physical retention, transmission of amplified sound to the ear canal, and control of the direct sound path between the ear canal and the air outside the head.

There is a wide variety of physical styles of both earmoulds and earshells. These styles vary in the extent of the concha and canal that they fill. These variations affect the appearance, acoustic performance, comfort, and security of retention of the hearing aid.

One unwanted consequence of a hearing aid can be an occlusion effect, in which the aid wearer's own voice is excessively amplified by bone-conducted sound. For many hearing aid fittings, vent selection is a careful juggle between choosing a vent that is big enough to avoid an unacceptable occlusion effect, but not so big that it causes feedback oscillations, or limits the ability to achieve sufficient low-frequency gain and maximum output. For patients with mild or moderate hearing loss, the choice will often be an extremely open fitting comprising a BTE connected to thin tubing, or a wire connection for a RITE style, terminating in a pre-formed, flexible, perforated, dome-shaped canal fitting.

For any ear fitting with a vent or other direct path to the outside air, the speech range of frequencies can be subdivided into the vent-transmitted frequency range, the amplified frequency range, and the mixed frequency range that is intermediate to these. Hearing aids perform very differently in each of these ranges.

The shape of the sound bore that connects the receiver to the ear canal affects the high-frequency gain and output of hearing aids. Sound bores that widen as they progress inwards (horns) increase the high-frequency output. Conversely, those that narrow (constrictions), whether by design or as a consequence of poor construction technique, decrease the high-frequency output. Horns have to exceed a certain length if they are to be effective within the frequency range of the hearing aid.

Dampers are used within the sound bore to smooth peaks in the gain-frequency response. Careful choice of the placement and resistance of the damper can also control the mid-frequency slope of the response.

The key to a well-fitting earmould is an accurate ear impression. This requires an appropriate material (medium viscosity silicone is good for most purposes), a canal block positioned sufficiently deeply in the canal, and smooth injection of the impression material.

Tighter earmoulds or shells that reduce leakage of sound from the ear canal can be achieved by a variety of techniques. These techniques include taking an impression with the patient's jaw open, patting down the impression material before it sets, using viscous impression material, and building up the impression in the patient's ear.

Earmoulds are made from a variety of materials. The most important difference between materials is hardness. Soft materials provide a better seal to the ear, but they deteriorate more rapidly, can be more difficult to insert, and are more difficult to modify and repair. Earmoulds and earshells are routinely constructed by computer-aided manufacture in which lasers guide the "printing" of plastic based on a scanned image of the ear impression.

Compression Systems in Hearing Aids

The major role of compression is to decrease the range of sound levels in the environment to better match the dynamic range of a hearing-impaired person. The compressor that achieves this reduction may be most active at low, mid, or high sound levels. More commonly, it will vary its gain across a

wide range of sound levels, in which case it is known as a wide dynamic range compressor. Compressors can be designed to react to a change in input levels within a few thousandths of a second, or their response can be made so gradual that they take many tens of seconds to fully react. These different compression speeds are best suited to different types of people.

The degree to which a compressor finally reacts as input level changes is best depicted on an input-output diagram or on an input-gain diagram. The compression threshold, which is the input level above which the compressor causes the gain to vary, is clearly visible on such diagrams. The compression ratio, which describes the variation in output level that corresponds to any variation in input level, is related to the slope of the lines on these diagrams.

Simple compression systems can be classified as input-controlled, which means that the compressor is controlled by a signal prior to the hearing aid's volume control, or as output-controlled, which means that the compressor is controlled by a signal subsequent to the volume control. This classification is irrelevant for hearing aids with no volume control and inadequate for hearing aids with multiple, sequential compressors.

Compression systems have been used in hearing aids to achieve the following more specific aims, each of which requires different compression parameters. Output-controlled compression limiting can prevent the hearing aid ever causing loudness discomfort, or the signal being peak clipped. Fast-acting compression with a low compression threshold can be used to increase the audibility of the softer syllables of speech, whereas slow-acting compression will leave the relative intensities unchanged, but will alter the overall level of a speech signal. Compression applied with a medium compression threshold will make hearing aids more comfortable to wear in noisy places, without any of the advantages or disadvantages that occur when lower level sounds are compressed. Multichannel compression can be used to enable a hearing-impaired person to perceive sounds with the same loudness that would be perceived by a normal-hearing person listening to the same sounds. Alternatively, it can be used to maximize intelligibility, while making the overall loudness of sounds normal (rather than the loudness at each frequency). Compression can be used to decrease the disturbing effects of background noise by reducing gain most in those frequency regions where the SNR is poorest. Gain reduction of this type increases listening comfort and with some unusual noises may also increase intelligibility. Finally, compression can be applied by using the combination of compression parameters that patients are believed to prefer, irrespective of whether there is a theoretical rationale guiding the application. Although these rationales are different, they have various aspects in common. Furthermore, many of them can be combined within a single hearing aid.

Despite the complexity, the benefits of compression can be summarized simply, but accurately, as follows. Compression can make low-level speech more intelligible, by increasing gain, and hence audibility. Compression can make high-level sounds more comfortable and less distorted. In mid-level environments, compression offers little advantage relative to a well-fitted linear aid. Once the input level varies from this, of course, the advantages of compression become evident. Its major disadvantages are a greater likelihood of feedback oscillation, and excessive amplification of unwanted lower level background noises.

Directional Microphones and Arrays

Other than the use of a remote microphone located near the source, directional microphones (which work by sensing sound at two or more locations in space) are the most effective way to improve intelligibility in noisy environments.

Directivity is most commonly achieved in hearing aids with first-order subtractive directional microphones, in which the output of one omni-directional microphone is delayed and subtracted from the output of the other. This internal delay, relative to the physical spacing between the two microphone sound ports, largely determines the polar sensitivity pattern of these microphones. The head itself also affects the polar pattern.

These subtractive directional microphones inherently cause a low-frequency cut in the frequency response, for which the hearing aid signal processing often compensates by a low-boost characteristic, but which also causes greater internal noise in the hearing aid. Split-band directivity, which combines a directional response for the high frequencies with an omni-directional response for the low frequencies, avoids this problem, but of course provides no noise reduction for the low frequencies. Irrespective of the frequency range over which the microphone is directional, the complete hearing aid fitting will have directivity only over the frequency range for which the gain of the amplified sound path exceeds that of the vent sound path. In open fittings, this will likely be only half the speech frequency range. Whether achieved by split-band processing, or by the acoustics of an open fitting, the resulting pattern of high-frequency directivity and low-frequency omni-directional processing simulates the directivity pattern of normal hearing.

Additive directional arrays create directivity by adding together the output of two or more omni-directional microphones. They do not create additional internal noise. To be effective, however, the microphones have to be separated by distances larger than a quarter of the sound's wavelength. They are therefore less suitable for hearing aids, but are suitable for accessories such as hand-held microphones.

These simple fixed subtractive and additive arrays have a fixed pattern of sensitivity versus direction of the incoming sound. Adaptive arrays, by contrast, have directional patterns that vary depending on the location, relative to the aid wearer, of background noises. Adaptive arrays automatically alter the way they combine the signals picked up by two or more microphones so as to have minimum sensitivity for sounds coming from the direction of dominant nearby noise sources. The multiple microphones that provide the input signals can be mounted on one side of the head or on both sides of the head.

The most sophisticated directional microphone arrays apply complex, frequency-dependent, adaptive weights to the outputs of each omni-directional microphone before combining them. Like all directional microphone arrays, complex adaptive arrays work most effectively in situations where there is a low level of reverberant sound.

Directional microphones are effective when either the target speech or the dominant (rearward) noise source(s) are closer to the aid wearer than the room's critical distance (at which the reverberant and direct sound fields have equal intensity). In the special case of a close frontal talker and many distant noise sources, the improvement in SNR will approximate the directivity index of the hearing aid averaged across frequency.

The disadvantages of directional microphones include insensitivity to wanted sounds from the sides or rear, increased internal noise if used in quiet places, reduced localization accuracy if the two hearing aids act in an uncoordinated manner, and increased sensitivity to wind noise. These disadvantages can be minimized by intelligent switching (automatically or manually) between directional and omni-directional modes, on the basis of noise levels and apparent SNR at the output of the omni and directional microphones.

All hearing aid wearers are candidates for directional microphones because all hearing aid wearers need a better SNR than people with normal hearing. Adaptive directional arrays that combine (via a wireless link or a cable) the outputs of microphones on both sides of the head produce a super-directional response that should enable people with mild hearing loss to hear better than people with normal hearing in many social situations.

Advanced Signal Processing Schemes

Adaptive noise reduction schemes, such as Wiener Filtering and Spectral Subtraction, progressively decrease the gain within each frequency region as the SNR deteriorates. Although they generally improve sound comfort and the overall SNR, these schemes do not change the SNR in any narrow frequency band. Consequently, they do not generally improve intelligibility. Other types of noise reduction include wind noise reduction, achieved by a low-frequency cut, and transient or impulsive noise reduction, achieved by limiting the rate at which the waveform changes.

Feedback oscillation can be made less likely by several electronic means. One simple technique is to decrease the gain only for those frequencies and input levels at which oscillation is likely. A second technique is to modify the phase response of the hearing aid so that the phase shift needed for oscillation does not occur at any frequency for which there is enough gain to cause feedback oscillation. A third technique involves adding a controlled internal negative feedback path that continuously adapts to maintain the gain and phase response needed to cancel the accidental leakage around the earmould or shell. A final technique involves making the output frequency different from the input frequency. Often, a combination of these techniques is used.

High-frequency components of speech can be made more audible by lowering their frequency. This can be achieved by transposition: moving sections of the spectrum to lower frequencies and superimposing them on the spectrum already in the lower frequency range. Alternatively, frequency compression is used to compress a wide frequency range into a narrower (and lower) one. While both frequency transposition and frequency compression can guarantee audibility of high-frequency sounds, they do not necessarily guarantee better intelligibility, as the speech components shifted down in frequency may interfere with perception of the speech components that were originally dominant in this lower frequency range. The range of possible frequency lowering methods, frequency transformation maps, and gain characteristics is large. Finding the best combination is made more difficult by the time it takes people to adapt to a highly altered spectrum and by our present uncertainty over how best to evaluate success.

There are various theoretically appealing methods for enhancing features of speech that have been tried in research experiments. These include exaggerating the peaks and troughs in the spectrum of a speech sound, increasing the amount of amplification whenever a consonant occurs, increasing the amplitude at the onset of sounds, lengthening and shortening the duration of particular sounds,

simplifying speech down to a few rapidly changing pure tones, and resynthesizing clean speech based on the output of an automated speech recognizer. On the evidence available so far, however, none of these techniques will produce a worthwhile increase in intelligibility compared to conventional amplification, so there is as yet little motivation to include the processing within commercial hearing aids.

Various other signal processing algorithms have already been implemented in commercial hearing aids, or could readily be implemented. Reverberant energy that does not overlap other speech sounds can be removed, giving a crisper sound quality. Hearing aids can automatically categorize the listening environment they are in, and select amplification characteristics that have been pre-programmed into the hearing aid for each type of environment. Their data-logging systems can record how often each environment is encountered, and how the user adjusts the hearing aids in each environment. Trainable hearing aids can learn from the adjustments the aid wearer makes, and infer how the aid wearer likes the hearing aid to be adjusted as the acoustics of the listening situation vary. Fine tuning is therefore carried out by the hearing aid and the client together, rather than by the clinician. Active occlusion reduction processing enables a hearing aid to sound like an open-canal hearing aid, despite the ear canal being completely blocked. In addition to cancelling the occlusion-induced sound, active occlusion reduction cancels the vent-transmitted sound, enabling directional microphones to work over the entire frequency range, increasing their efficacy.

Assessing Candidacy for Hearing Aids

Although the decision to try hearing aids is ultimately made by the patient, many patients will be in doubt as to whether they should acquire hearing aids and so will look to the clinician for a recommendation. This recommendation must take into account many factors other than pure tone thresholds.

Initial motivation to obtain hearing aids has been shown to be a key determinant of whether patients continue to use them. This motivation reflects the balance of all the advantages a patient expects hearing aids will provide offset by all the expected disadvantages, irrespective of whether all these positive and negative expectations are realistic. The advantages expected by the patient are affected by the degree of disability they feel they have. Disability includes how much difficulty the person has hearing in various situations, referred to as activity limitation, and the extent to which a person is unable to participate in activities because of the hearing loss, referred to as participation restriction. The advantages and disadvantages expected by the patient are affected by what the patient has been told about hearing aids by others. Disadvantages potentially include the impact on a patient's self-image of wearing hearing aids. The clinician must attempt to discover a patient's expectations and modify those that are unrealistically low or unrealistically high. Although hearing aids help in quiet and in noise, they help much more in quiet, so hearing aids are more likely to be valued and used if the patient needs help hearing in quiet places.

When a clinician encounters a hearing-impaired patient who does not want hearing aids, the clinician should find out whether this is because the patient is not aware of the loss and/or the difficulty that he has compared to others, or because the patient, although acknowledging the loss, does not wish to wear hearing aids. If the latter is true, the patient's reasons should also be discovered.

Difficulty managing a hearing aid can greatly affect use, so the clinician must consider likely manipulation difficulties when determining candidacy and aid type. People with tinnitus often find that hearing aid use diminishes their problems, so tinnitus positively affects candidacy. The presence of central processing disorders and extreme old age can both affect candidacy, but not in a manner sufficiently predictable to affect the clinician's recommendation. People who are not worried that hearing aids will stigmatize them are more likely to acquire them, and people who more readily accept the presence of noise while listening to speech are more likely to use them. Several personality characteristics also make hearing aid acquisition, use, and/or benefit more likely.

People with a severe to profound hearing loss are likely to receive more benefit from cochlear implants than from hearing aids. The most useful indicator of which device will be better for them is the speech score they receive for well fitted hearing aids after some years of becoming accustomed to them. For infants, this is not possible so the decision to implant has to be based primarily on aided or unaided hearing thresholds (as well as requiring no medical or psychological contra-indications). Cochlear implants and hearing aids generally provide complementary cues, whether they are worn in the same ear, or in opposite ears.

Vibrotactile or electrotactile aids are a worthwhile alternative for those with too much hearing loss to receive useful auditory stimulation from hearing aids, but who do not wish to receive a cochlear implant, or for whom a cochlear implant is not suitable on medical or psychological grounds. Training in integrating the tactile information with visual information is essential.

Hearing aids should not be withheld just because speech scores obtained under headphones fall below some arbitrarily determined criterion. There are, however, several audiological/medical indications that should cause hearing aid fitting to be delayed until the cause of the problems has been resolved.

A clinician therefore has to consider a large number of factors that may affect candidacy for hearing aids, none of which has such a strong effect that the remaining factors can be ignored.

Prescribing Hearing Aid Amplification

Amplification can be prescribed using a formula that links some characteristics of a person to the target amplification characteristics. Prescription formulae most commonly used are based on hearing thresholds, but some are based on supra-threshold loudness judgments.

Well-known procedures for linear hearing aids include POGO, NAL, and DSL. For all of these, gain can be prescribed based on hearing thresholds alone. These formulae all contain variations of the half-gain rule, but the variations are so different that the resulting prescriptions differ greatly, especially for people with a sloping hearing loss.

For non-linear hearing aids, all available prescription procedures include some aspect of normalizing the loudness of supra-threshold sounds. Several procedures (LGOB, IHAFF, DSLm[i/o] curvilinear, CAMREST and FIG6) aim to normalize loudness at all frequencies, at least for sounds with levels above the compression threshold of the hearing aid. Other procedures vary from loudness normalization in some way. ScalAdapt decreases the loudness of low-frequency sounds; CAM2 and NAL-NL2 normalize only the overall loudness. CAM2 aims to equalize the contributions that different frequency regions make to loudness, whereas NAL-NL2 aims for the sensation levels across

frequency that will maximize calculated speech intelligibility. As each of the formulae has been revised, their prescriptions have become more similar to each other, but marked differences still occur.

There are some issues related to prescription that are not yet resolved, although there is considerable information available about each issue. How much do patients' preferences and performance with hearing aids change following weeks, months, or years of experience with amplified sound? How loud (a perception, not a physical quantity) do patients like sound to be?

Should tests of dead regions in the cochlea routinely be conducted? How severe does hearing loss need to be before it is considered unaidable? As signal level decreases, down to how low a level should gain keep increasing? How accurately must prescription targets be met? What is the best combination of fast and slow compression?

Maximum output (OSPL90) has to be prescribed so that loudness discomfort is prevented, but so that enough loudness can be obtained without the hearing aid becoming excessively saturated. In many procedures, the target OSPL90 is assumed to just equal LDL, in others it is predicted from threshold, in which case it may fall above or below an individual patient's LDL as measured in the clinic. For patients with mild to severe hearing loss, an acceptable sound quality is more likely if compression limiting controls maximum output than if peak clipping controls maximum output. Many patients with a profound loss, however, will benefit from the additional SPL that is achievable with a peak clipper.

People with conductive and mixed hearing loss require greater gain and OSPL90 than people with sensorineural loss of the same degree. For a variety of reasons, the gain needed to compensate for a conductive loss seems to be less than the amount of attenuation that the conductive loss causes in the middle ear. Consequently, the same is true of OSPL90.

Multi-memory hearing aids can have a different prescription for each memory. These alternatives can be prescribed as variations from the baseline response prescribed for the first memory. The variations are designed to optimize specific listening criteria or for listening to different types of signals, such as music. People who wear their hearing aids in many environments, have more than 55 dB high-frequency hearing loss, and require more than 0 dB low-frequency gain, are most likely to benefit from multiple memories.

Neither gain nor OSPL90 should be any higher than is necessary for a patient. Otherwise, a hearing aid may increase hearing loss because of the resulting high-level exposure to sound. The risk of temporary or permanent noise-induced loss is greatest for patients with a profound loss, and can be minimized by using non-linear amplification.

Selecting, Adjusting and Verifying Hearing Aids

The first decision to be made when a clinician and patient select a hearing aid is whether a CIC, ITC, ITE, BTE-RITE, BTE-RITA (with standard tubing and earmould, or thin tube and earmould or instant-fit dome), spectacle, body style, or a sub-variety of any of these, would be most suitable. For each style there are advantages relating to ease of insertion, ease of control manipulation, visibility, amount of gain, sensitivity to wind noise, directivity, reliability, telephone compatibility, ease of cleaning,

avoidance of occlusion and feedback, ability to assess and fit in the same appointment, and cost. The weight given to each factor will vary greatly from patient to patient.

The need for specific features, such as a volume control, a telecoil and switch, a direct audio input, and a directional microphone must be determined on an individual basis. These needs will also influence the style of hearing aid selected. BTEs have more advantages than the other styles for a majority of patients.

Next, signal-processing options appropriate to the needs of the patient must be selected.

Compression limiting is a more appropriate form of limiting than peak clipping if it can provide a high enough maximum output. In addition to compression limiting, a low compression ratio, active over a wide range of input levels, is appropriate for most patients. This low-ratio compression will provide advantages whether it is single- or multichannel, and whether it is fast or slow acting. Multichannel compression will provide greater speech intelligibility and/or comfort for patients with moderately or steeply sloping hearing loss, and the multichannel structure facilitates other features such as adaptive noise suppression and feedback suppression. The comfort advantages of adaptive noise reduction are greatest for patients who wear their hearing aids in a range of environments and who also require amplification across a wide range of frequencies. These same considerations apply to multi-memory hearing aids, the only difference being that the patient, rather than the hearing aid, chooses the response variations. Feedback cancellation is most beneficial for patients with a severe or profound hearing loss, patients with a severe loss in the high frequencies but near-normal hearing in the low frequencies, any patients fit with open canal devices, and any patient who wishes to use the telephone without using telecoil input. This combination makes it useful for nearly every client. Frequency lowering is advantageous for some patients though it is not yet possible to predict which patients will benefit. Trainability enables patients to take responsibility for fine-tuning their hearing aids.

Hearing aid fitting software provides a first approximation to the prescribed gain-frequency response target. The software must appropriately allow for the acoustic configuration of the earmould shell or dome fitting. The approximation can be made even more accurate by incorporating the individual patient's real-ear to coupler difference (RECD) in the prescription. This increased accuracy in the pre-calculation is probably only worthwhile for hearing aids intended for infants, where measurement of the final real-ear gain is difficult.

Any signal-processing scheme that requires adjustment for each patient must also be supported by an appropriate prescription method. Measurement of real-ear gain is necessary unless the hearing aid has been adjusted in the coupler using the individual's (or at least an age-appropriate) RECD.

Because it is not possible to prescribe OSPL90 with complete precision, the suitability of maximum output should be subjectively evaluated before the patient leaves the clinic. A variety of intense sounds should be presented to the patient to ensure that the hearing aid does not make sounds uncomfortably loud. Maximum output must, however, be great enough for the patient to experience intense sounds as being loud. This can be assessed by presenting speech signals at a high level and asking the patient to report their loudness.

Problem Solving and Fine Tuning

Many hearing aid fittings need to be fine-tuned, either electronically or physically, after the patient has had a week or two to try the hearing aids. When a patient has trouble managing hearing aids (inserting, removing, using the controls, changing the battery), re-instructing the patient may solve the problem. If not, the hearing aid should be physically modified, or if necessary, a different style chosen. Physical modification will also be necessary when a patient is suffering discomfort from the earmould, shell or case, or when the hearing aid works its way out of the ear.

Feedback oscillation has several potential solutions: reducing gain at selected frequencies; reducing the vent size; making a tighter earmould or shell; or changing the hearing aid to one that has more effective feedback canceling and management algorithms.

Complaints about the patient's own-voice quality are particularly common. The most common cause is physical blocking of the ear canal, so the best cure is to add a vent, or increase the size of an existing vent, including using an open-fitting. Where feedback oscillation precludes that, the earmould or shell can be remade with the canal stalk extended down to the bony canal, preferably using a soft material. Own-voice problems are sometimes caused, and cured, by electronic variation of the gain-frequency response for high-level sounds.

Complaints about the tonal quality of amplified sounds are fixed by changing the balance of low-, mid-, and high-frequency gain. The hard part is knowing when to ask the patient to persevere with a gain-frequency response in the expectation that it will eventually become the preferred response, and confer maximum benefit to the patient.

When a patient complains about the clarity or loudness of speech, or the loudness of background noise, he/she must be questioned particularly carefully so that the acoustic characteristics of the sounds causing the problems can be identified. The clinician's first aim is to identify whether it is the gain for low or high frequencies, and the gain for low, mid, or high levels, that should be adjusted. Only then can the appropriate hearing aid controls be adjusted.

In those cases where it is not clear which control should be adjusted, or by how much it should be adjusted, a systematic fine-tuning can be performed using one of two general methods. The first of these is paired comparisons, in which the patient is asked to choose between two amplification characteristics presented in quick succession. Multiple characteristics can be compared by arranging them in pairs. Paired comparisons can be used to adaptively fine-tune a hearing aid control if the settings compared in each trial are based on the patient's preference in the preceding trial.

The second general method for fine-tuning relies on the patient making an absolute rating of sound quality. The best amplification characteristic (out of those compared) is simply the characteristic that is given the highest rating by the patient. The absolute rating method can also be used to adaptively alter a chosen hearing aid control. This is achieved by deciding on a target rating (e.g. just right) and adjusting a control in the direction indicated by the patient's rating (e.g. too shrill, or too dull).

The paired comparisons and absolute rating methods are best carried out while the patient listens to continuous discourse speech material, or other sounds they are complaining about. Depending on the complaint being investigated, this can be supplemented with recordings of commonly encountered background noises. The paired comparisons method is more sensitive when the

differences between the conditions are small. Fine-tuning is usually carried out only for patients dissatisfied with the prescribed response, but can be used for all patients if desired.

Patient Education and Counseling for Hearing Aid Wearers

People with a hearing impairment benefit from patient education and may benefit from communication training and counseling. These activities may be aimed at giving patients information about their hearing loss, developing skills needed to operate and care for their new hearing aids, improving listening skills, or changing patients' beliefs, feelings and behaviour relating to their hearing and communication. Providing appropriate education and counseling increases the likelihood that hearing aids will be fully used and that residual communication difficulties will be minimized.

It is difficult to help patients understand the variety of hearing aid styles and performance features that may be suitable for them. The benefits and cost implications of each (including ongoing service costs, warranty, and trial periods) have to be presented in a suitably simple manner.

Once they start using their hearing aids, first-time hearing aid users experience a new world of amplified sound, and may benefit from guidance about how to gradually increase their range of listening experiences. The aim is to provide them with the best experiences first, and to avoid having them become overwhelmed by sound. Patients need to know that their brains may take some time to adapt to hearing parts of speech, and other sounds around them, that they have not heard for some time.

A major part of educating the new hearing aid user has nothing to do with hearing aids! A wide range of hearing tactics and strategies can help the hearing-impaired person understand more in difficult listening situations. The first group of hearing strategies requires the listener to look carefully at the talker and the surroundings. The second group requires the listener to alter the communication pattern in some way. The final group requires the listener to manipulate the environment to remove or minimize sources of difficulty. Patients will benefit if family members and/or other frequent conversation partners participate in education sessions on these topics.

Patients will more easily appreciate and learn this material if it can be taught in a patient-centred, individual problem-solving method, rather than as a set of rules disconnected from their everyday lives. Communication training comprises training in the use of these hearing strategies, plus practice in listening to speech (synthetic training) or to the basic sounds from which speech is built (analytic training), especially in difficult listening conditions. Increasingly, communication training is being provided in packages that patients can use on their computer or DVD at home. Patients should be advised about protecting their remaining hearing, and be made aware of where they can obtain support (from peer groups or other professionals) beyond that which the clinician can provide.

Hearing aids do not provide an adequate solution to all hearing problems, so patients must be made aware of other assistive listening devices that may help them. Clinicians should be aware that different people learn in different ways. Consequently, the same material should be taught in different ways to different patients, and clinicians should develop the flexibility needed to accomplish this.

Clinicians must be flexible regarding how and when they present information and carry out other procedures. It is, nonetheless, useful to have in mind a standard program from which variations can be made as required.

Assessing the Outcomes of Hearing Rehabilitation

Clients and clinicians both benefit when the outcomes of the rehabilitation process (i.e. changes in the patients' lives) are measured in some way. Systematic measurement of outcomes can help clinicians learn which of their practices, procedures, and devices are achieving the intended aims. Some measures can also help determine how the rehabilitation program for individual patients should be structured and when it should be ended.

Outcome assessment can be based on an objective speech recognition test (the results of which depend hugely on the measurement conditions), or on a subjective self-report and/or the report of a significant other person. Speech test scores show the increase in the ability to understand speech in specific situations, whereas self-report measures more generally reflect the patient's views about the impact of rehabilitation.

Many self-report measures have sub-scales so that outcomes can be separately assessed for different listening environments. Outcome measures can assess the domains of benefit, defined as a reduction in disability (comprising activity limitation and participation restriction), device usage, listening effort, quality of life, or the satisfaction that the patient feels.

Self-report measures that assess benefit can be grouped into various classes. First, patients can be asked to make a direct assessment of the benefit of rehabilitation. Alternatively, patients' views of their disability can be assessed both before and after the rehabilitation program. The change in score provides a measure of the effects of rehabilitation. Measures obtained both before and after rehabilitation provide a more complete view of disability status and change. These difference measures probably assess change less accurately than those that directly assess benefit because they involve subtracting two scores.

The second way in which self-report measures differ from each other is the extent to which the items are the same for all patients or are determined individually for each patient. Results can more easily be compared across patients if a standard set of items is used for all patients. When the items are individually selected for each patient, however, the questionnaires become shorter and can more easily be incorporated within interviews with the patient. They are also more relevant to each patient.

There are thus four types of self-report measures: standard questionnaires that directly assess benefit (e.g. HAPI); standard questionnaires that compare disability before and after rehabilitation (e.g. HHIE, APHAB); individualized questionnaires that directly assess benefit (e.g. COSI); and individualized questionnaires that compare disability before and after rehabilitation (e.g. GAS).

Self-report measures also commonly assess hearing aid usage (which can now also be measured objectively with data logging) and are the only viable way to assess satisfaction. Some measures contain questions that address only one domain (benefit, use, or satisfaction) whereas others address more than one domain. One comprehensive questionnaire (GHABP) addresses all three dimensions, contains standard and individualized measures, and assesses benefit both directly and

by comparing disability before and after rehabilitation. A very simple and widely used questionnaire that assesses several domains with a single question each is the International Outcomes Inventory for Hearing Aids (IOI-HA).

Some questionnaires are designed to assess problems experienced with the hearing aid, although freedom from problems with the hearing aid is more properly viewed as a means to an end rather than a life-changing outcome.

While outcomes can be assessed any time after hearing aid fitting, the extent of benefit does not appear to stabilize until about 6 weeks after fitting. Hearing loss is associated with a decrease in many aspects of quality of life (such as increased depression) and use of hearing aids is associated with general improvements in health and quality of life. Causal relationships between these quantities are difficult to establish, however. Generic measures of health outcome are not efficient means by which a clinician can assess the outcomes of rehabilitation.

Binaural and Bilateral Considerations in Hearing Aid Fitting

Sensing sounds in two ears (binaural hearing) makes it possible for a person to locate the source of sounds and increases speech intelligibility in noisy situations. Wearing two hearing aids (a bilateral fitting) instead of one hearing aid (a unilateral fitting) increases the range of sound levels for which binaural hearing is possible. Bilateral fitting is thus more important when hearing loss is severe than when it is mild or moderate.

Accurate horizontal localization is possible because sounds reaching the two ears differ in level and in arrival time, and hence in phase. These cues are also present, but altered, when people wear hearing aids. Most hearing-impaired people, once they become used to the effect of their hearing aids on these cues, can localize sounds accurately to the left and right in the horizontal plane. Vertical localization and front-back localization, which are based on very high-frequency cues created by the pinna, are extremely adversely affected by hearing loss and are not significantly improved by hearing aids.

When speech and noise arrive from different directions, head diffraction causes the signal-to-noise ratio to be greater at one ear than at the other. Further, the auditory system can combine the different mixtures of speech and noise arriving at each ear to effectively remove some of the noise. This ability is known as binaural squelch. Even presenting identical sounds to the two ears provides a small improvement in speech intelligibility over listening with one ear, a phenomenon known as binaural redundancy.

Wearing a second hearing aid will improve speech intelligibility in noise whenever it causes speech to become audible in the previously unaided ear. Achieving audibility of speech in both ears is a prerequisite to attending to the ear with the better signal-to-noise ratio, and to benefitting from binaural squelch and binaural redundancy. Bilateral fitting of hearing aids has several other advantages. These include improved sound quality, suppression of tinnitus in both ears, and greater convenience if one hearing aid breaks down or when one battery dies. A bilateral fitting may help prevent a problem sometimes associated with unilateral fittings: a unilateral fitting can lead to decreased speech processing ability in the unaided ear if this ear is deprived of auditory stimulation for too long, a phenomenon referred to as late-onset auditory deprivation.

The advantages of bilateral fittings also apply to patients with asymmetrical hearing thresholds. If such patients must receive a unilateral fitting, it may be generally advisable to fit the ear with thresholds closest to about 60 dB HL.

Bilateral fittings also have disadvantages: they cost more, are more susceptible to wind noise, and are more difficult for some elderly people to manage. Also, some people regard two hearing aids as an indication of severe hearing loss, and do not wish to be perceived in this way. For some people, binaural interference causes speech identification ability to be better when unilaterally aided than when bilaterally aided. The causes of this interference may lie in differences between the two cochleae, differences between the two hemispheres of the cortex, or distortions in transfer of information from one hemisphere of the cortex to the other.

Because of the variability associated with speech intelligibility testing, conditions have to be chosen carefully to reliably demonstrate bilateral advantage or detect binaural interference on an individual patient. To best demonstrate bilateral advantage, loudspeaker positions for speech and noise should be chosen to maximize the effects of head diffraction and binaural squelch. To best detect binaural interference, speech and noise should emanate from a single, frontal loudspeaker, so that the effects of head diffraction and binaural squelch are minimized. In either case, speech tests with steep performance-intensity functions should be used. A method for predicting whether individual patients will benefit more from a unilateral or bilateral fitting is urgently needed.

Special Hearing Aid Issues for Children

When a child is born with a hearing loss, early provision of hearing aids is essential if he or she is to learn to speak and listen with the greatest possible proficiency. Hearing aids should be provided by six months of age. If cochlear implants are a better option, these should be implanted by 12 months of age. Children with bilateral loss should receive bilateral hearing aids. There is some uncertainty over optimal treatment for children with unilateral loss, mild loss, or auditory neuropathy.

For the hearing aids to be optimally adjusted, frequency-specific hearing thresholds must be determined separately for each ear. No matter what type of transducer is used, the small size of a baby's ear complicates the interpretation of hearing threshold. This difficulty is overcome either by expressing threshold in dB SPL in the ear canal, or by expressing it as equivalent adult hearing threshold in dB HL.

BTE hearing aids are most likely to be provided, in conjunction with soft earmoulds, until the child is at least 8 years old (and possibly much older). The hearing aid should contain features that will enable the child to receive the best possible signal. This is likely to include an audio input socket and/or telecoil, and/or internal wireless receiver, so that there is some means to receive wireless transmission. Ideally, the wireless device should be able to automatically attenuate the local microphone whenever the person wearing the transmitter talks.

To communicate effectively, normal-hearing children learning language need a better signal-to-noise ratio than do adults. They also understand speech less well than adults at very low sensation levels. These observations may lie behind the empirical finding that hearing-impaired children prefer more gain than adults with the same hearing loss. Compared to adults, they almost certainly do not need

any more real-ear gain for high-level sounds, they probably prefer more gain for medium level sounds, and they almost certainly need more gain for low-level sounds.

There is an even greater need for wide dynamic range compression in hearing aids for children too young to manipulate the volume control than there is for adults. Similarly, infants have an even greater need than adults for directional microphones and adaptive noise reduction systems. These algorithms also have potential disadvantages, but should be provided if the clinician has confidence in the automated manner in which the hearing aid selects them.

To achieve a certain real-ear gain, young children need less coupler gain than do adults, because children have smaller ear canals. An efficient way to allow for small ear canals is to measure real-ear to coupler difference before prescribing the hearing aid, and to calculate the coupler gain that will result in the target rear ear aided gain. A faster but less accurate way is to use age-appropriate values of real-ear to coupler difference.

The maximum output that has been prescribed should be evaluated by observing the child when intense sounds are made and, for those over approximately six years of age, by assessing the loudness of these sounds.

Hearing aid fittings can be evaluated by speech testing (for those over three years of age), paired-comparison preference testing (for those over six years of age), and subjective reporting by the child, the parents, or the teachers (whether informally or using systematic methods like PEACH and MAIS). The audibility of speech can be estimated by calculating the articulation index (also known as speech intelligibility index) or assessed by measuring the presence, latency and perhaps morphology of the cortical responses elicited by speech sounds. The availability of speech to the child can be indirectly assessed by measuring the child's language development.

Effective amplification for young children is not possible without the support and understanding of parents. The audiologist must therefore inform and support the parents in a variety of ways. One way to provide ongoing habilitation is to base the service activities around goals determined jointly by the audiologist and the parents (and by the child when old enough).

Part of the information provided to parents includes safety aspects of amplification and hearing loss. Hazards include battery, earmould or hearing aid ingestion, excessive exposure to noise, physical impact, and failure to detect warning signals if amplification is not functioning correctly.

CROS, Bone-conduction and Implanted Hearing Aids

In the CROS (Contralateral Routing of Signals) family of hearing aids, hearing aid components on opposite sides of the head are wirelessly linked. Basic CROS aids are most suitable for people with unilateral loss. CROS aids consists of a microphone on the side of the head with a deaf ear, combined with an amplifier, receiver and open earmould or shell on the side with a normal-hearing ear. Adding a microphone to the side of the better ear converts it to a BICROS hearing aid, which is suitable for patients with loss in both ears. A transcranial CROS has all the components in one ear, but sends a signal across the head by bone conduction. CROS hearing aids must be carefully fitted to ensure the aid wearer receives, in a single cochlea, an appropriate balance of sounds reaching the two sides of the head.

Bone-conduction hearing aids output a mechanical vibration instead of an air-borne sound wave. They are most suited to people who, for medical or anatomical reasons, cannot wear a hearing aid that occludes the ear in any way, or for those who have a large conductive loss in either ear. For patients with normal external and middle ears, bone-conduction hearing aids cannot stimulate the cochlea as effectively as do air-conduction hearing aids because of the relative inefficiency of the bone conduction pathway. For patients with maximal conductive hearing losses, whether or not there is also a sensorineural loss, bone-conduction hearing aids can stimulate the cochlea as strongly as air-conduction hearing aids. Prescriptions for air-conduction hearing aids can be converted into bone-conduction prescriptions by using available standards for the thresholds of hearing for air- and bone-conducted sound. Bone-conduction output is specified in terms of output force level instead of output sound pressure level, and in terms of acousto-mechanical sensitivity instead of gain.

Disadvantages of non-implanted bone conduction hearing aids include their wearing comfort and the limited sensation level they can provide. A commonly used form of bone-conduction hearing aid is the bone-anchored hearing aid, in which the vibrations are transmitted to the skull via an embedded titanium screw, thereby increasing stimulation of the cochlea by about 15 dB compared to a bone conductor applied to the skin. Bone-anchored hearing aids have successfully been used for patients with unilateral or bilateral conductive or mixed loss. They are also routinely being fitted to people with unilateral sensorineural loss, referred to as single-sided sensorineural deafness. The output levels they can achieve make them suitable for people with cochlear loss up to about 45 dB HL for head-worn devices and up to about 60 dB HL for body-worn devices. Bone-anchored hearing aids can provide greater cochlear stimulation than air conduction hearing aids for patients with air-bone gaps greater than about 30 dB.

A variety of other middle-ear implants have been researched, and several have been approved for routine use. Middle-ear implants may have only the output transducers surgically implanted, or may be combined with implanted microphones and batteries to form completely implanted hearing aids. Four types of output transducers have been used: magnets enclosed by a coil that rely on the inertial mass of the magnet, magnets mounted on the middle-ear chain driven by a remote coil, and piezoelectric or electromagnetic stimulators anchored to the mastoid bone and vibrating the middle-ear chain. Three types of microphones have been used: external microphones, microphones implanted under the skin on the scalp or in the ear canal, and transducers that are driven by the vibration of the middle-ear chain.

Several implanted hearing aids are now commercially available. For some clients, particularly those with mixed hearing loss, middle-ear implants may have advantages related to freedom from occlusion, amplification gain and bandwidth, stimulation level, and invisibility of the device. Fully implanted devices have additional advantages arising from there being no external parts. Candidacy criteria are still developing.

Reference

The material of this chapter comprises the one-page synopses of each chapter of the book "Hearing Aids" (Dillon, 2012, Thieme, New York) . That book elaborates on all the information covered in this chapter, including citations of 2,000 references.