2008

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CICSD

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Hearing aids have undergone significant improvements since the days when ear trumpets and animal horns were used to transmit sounds to the ear. As digital technology came to the forefront in the field of telecommunications, its application expanded to the field of amplification. In 1996, digital signal processing (DSP) was introduced into hearing aid processing algorithms (Edwards, 2007).

Since then, advances in digital technology and the processing speed of the DSP chip have opened the door to expand and improve the available features in modern hearing aids. Even with these advances, however, hearing aid users continue to complain of difficulty hearing speech in noise and performance with telephones. According to the MarkeTrak V data, 25.3% of hearing aid users reported that they have a hard time listening when there is a lot of background noise present, and approximately 1% of users reported difficulty using the telephone (Kochkin, 2000).

ABSTRACT: Digital signal processing (DSP) has provided significant advances and improvements in hearing aid technology. These advances, and the increased processing speed that DSP offers, have resulted in improved and expanded features in current hearing aid technology. These advances have also served to expand the fitting range of amplification for patients who previously could not benefit from amplification. This article reviews the advances and supporting research in specific hearing aid features, devices, and assistive technology developments that allow a greater number of patients access to amplification. These features and devices include directional microphones, Bluetooth and other wireless technology, digital noise reduction, devices for patients with single-sided deafness, frequency transposition, self-learning and adaptation managers, and integrated real ear measures. Despite the changes in hearing aids created by DSP, limitations still exist. Additionally, peer-reviewed research is not yet available to support the reported benefits of some of these advances.

KEY WORDS: digital noise reduction, digital signal processing, directional microphones, frequency compression, frequency transposition

Recent developments in directional microphone technology and improvements in access to and use of wireless technology are improving speech recognition in noise, or the signal-to-noise ratio (SNR). Access to newer forms of wireless transmission and improvements in coupling this technology to hearing aids not only enhances patients' abilities to use telephones, but also allows patients to directly connect to a multitude of external devices. Noise reduction algorithms have also advanced to help improve hearing aid users' comfort and ease of listening in noise, and several algorithms have attempted to improve speech recognition in noise. In addition, several devices and features are now being used to assist individuals with hearing impairment who have not previously experienced benefit with hearing aids. There are also devices and/or features that make the fitting process easier for the audiologist and the patient. These newer devices include hearing aids for single-sided deafness (SSD), frequency transposition for patients exhibiting severe high-frequency hearing loss, self-learning features on hearing aids, and hearing aids that allow for integrated real ear measures.

DIRECTIONAL MICROPHONES

Of all of the advances in hearing aid technology in the last several years, perhaps the greatest has been the performance of directional microphones. The use of DSP in hearing aids has opened the door to many different types of

1092-5171/08/3502-0102

algorithms used in directional microphones (Preves & Banerjee, 2008). Before the introduction of DSP, analog hearing aids had the capability to use fixed directionality only. Many of these analog hearing aids either contained a dedicated directional microphone or had a push button that was used to switch the hearing aid from omnidirectional to directional microphone mode. When in the directional microphone mode, the direction where the sound was being attenuated—the null—remained the same (i.e., “fixed”) regardless of the direction of the sound source. With advances in digital technology, there are now many options for directionality, including automatic, automatic adaptive, multiband automatic adaptive, and, most recently, asymmetric directionality. Each of these options has benefits, but some also have limitations and may not prove to be as beneficial to the patient as touted by hearing aid manufacturers.

Listeners with even a mild hearing loss can have a 4-dB deficit in their SNR performance, and listeners with greater degrees of hearing loss will have even greater deficits (Killion, 1997). Directional microphones were developed in an attempt to improve SNR performance for the hearing aid user. The fixed directional microphone contains two sound ports and operates by acoustically delaying the signal entering the back microphone port and subtracting this from the signal entering the front port (Chung, 2004). This creates a null at an azimuth corresponding to the location where the microphone is least sensitive and can be plotted graphically on a polar pattern (Chung, 2004). Directional microphones can employ different types of polar patterns, some of which have multiple nulls. With the fixed directional microphone, these polar patterns are predetermined so the location of sound attenuation always remains the same. Typically, the null is fixed at 180°, creating a cardioid polar plot. Therefore, if the interfering sound is located directly behind the patient, this design acts to attenuate the input level to the hearing aid at the 180° null. If, however, the offending sound arrives from behind but not directly at 180°, then the microphone will be less effective in improving the SNR.

Several studies have reported the effectiveness of fixed directional microphones in improving the SNR for the hearing aid user. Bilsen, Soede, and Berkhout (1993) demonstrated that listeners with hearing impairment can achieve at least a 5-dB improvement in SNR by using fixed directional microphones. Gravel, Fausel, and Liskow (1999) found that children listening with dual microphones achieved a mean improvement of 4.7 dB in SNR when compared to the omnidirectional condition. Pumford, Seewald, Scollie, and Jenstad (2000); Ricketts and Dhar (1999); and Valente, Fabry, and Potts (1995) also found significant improvement in speech recognition scores in noise with a fixed directional microphone.

Although research has demonstrated that fixed directional microphones can improve hearing aid users’ speech recognition scores in noise, this feature has its limitations. Walden and Walden (2004) reported that most hearing aid users preferred the directional mode in the presence of background noise. However, many patients either do not know or do not remember when to appropriately switch their hearing aids to the directional mode. Consequently, they may report that they do not receive benefit from the directional microphone on their hearing aids. Cord, Surr, Walden, and Dyrlund (2004) conducted a study comparing the directional microphone performance of hearing aid users who reported a lack of success with directional microphones and hearing aid users who reported a benefit from directional microphones. The study did not report any difference in the directional advantage of the directional microphone between these two groups. This suggests that although directional microphones have been shown to be successful in the laboratory, there is no guarantee that this success will be achieved in real-life situations for all hearing aid users because of the difficulty that some people have in manipulating the hearing aid’s controls.

Automatic directional microphones were subsequently developed so that patients would not have to bother with manually changing the hearing aid program setting to the directional microphone mode. Automatic directional microphones use an algorithm in which the microphones switch automatically between omnidirectional and directional. Input level, signal location, and SNR are factors that contribute to determining when the microphones switch (Preves & Banerjee, 2008). The automatic microphone feature works well for those patients who do not want to be concerned with switching between omnidirectional and directional modes manually. However, automatic switching can be problematic for patients when the microphones switch but the patient does not prefer the switching, or if the switching takes place too rapidly and amplifies unwanted sounds such as a cough or a dog barking (Preves & Banerjee, 2008). The other limitation with automatic directional microphones is that the null is fixed when in the directional mode. Depending on the location of the noise source and the azimuth of the null in the microphone, there is the possibility that the noise source may not be maximally attenuated.

The next advancement in directional microphones addressed the limitation of fixed nulls, including instruments with automatic directional microphones. This advancement is termed the automatic adaptive directional microphone. Automatic adaptive directionality uses the principle of switching microphones in response to acoustic input by automatically switching to the directional mode. In addition, these microphones automatically adjust the polar plot in response to the azimuth of the noise arriving from the rear. For example, if the noise is arriving from the back, the microphone will automatically create a cardioid polar plot. If, however, the noise is arriving from the side, the directional microphone will automatically switch to a bidirectional polar design. Finally, if the noise arrives at an angle between the side and back, the directional microphone will automatically switch to a hypercardioid polar design (Preves & Banerjee, 2008). Bentler, Palmer, and Dittberner (2004) reported that when a noise source was moving, listeners with hearing impairment performed more similar to normal hearing individuals with an automatic adaptive directional microphone than with a fixed directional microphone. Ricketts and Henry (2002) demonstrated a significant improvement in speech recognition for both a fixed and an automatic adaptive directional microphone in the presence of background noise. However, the automatic adaptive directional microphone showed some additional advantages, especially when the noise was presented from the side and there was minimal reverberation. Blamey, Fiket, and Steele (2006) reported an adaptive advantage of 1 dB compared to a fixed hypercardioid polar design. Ricketts and Hornsby (2003) reported a directional benefit in rooms with low and moderate reverberation except when the speech source was moved to a distal field of 4.8 m. Furthermore, Ricketts, Hornsby, and Johnson (2005) later found that the advantage of using the automatic adaptive directional microphone was greatly reduced when there was a moderate level of uniform noise present.

The multichannel automatic adaptive directional microphone was the next advancement in directional microphones. This type of directional microphone automatically switches between omnidirectional and directional modes and incorporates different polar designs in the number of frequency channels available in the hearing aid. This feature is designed to account for the many different scenarios and types of noise sources that are present in the environment. For example, when someone is outside on a windy day, the hearing aid may be in omnidirectional mode in the lower frequency region and directional mode in the higher frequency region. Therefore, when the user is listening to speech in windy conditions, the noise will not be above the level of speech (Preves & Banerjee, 2008). Empirical data to support the effectiveness of this feature, however, are not yet available.

The most recent trend in directional microphones is the asymmetric directional microphone fitting, which uses a directional microphone in one ear (usually the right ear because of its dominance for speech processing) and an omnidirectional microphone in the other ear. With this feature, the hearing aid user has the option to select which signal to focus on without the limitations of a traditional directional microphone system. The ear with the directional microphone will hear sounds coming from the front, and the ear with the omnidirectional microphone will hear sounds coming from every direction. In theory, this can reduce the possibility of missing sounds that are not coming from directly in front of the hearing aid user. In other words, if there is a signal coming from the side or the rear instead of the front of the hearing aid user, the ear with the omnidirectional microphone will be able to pick up the amplified signal and the hearing aid user can then turn to the signal for more optimal listening. This feature could be helpful for a student in a classroom environment. In this case, with one hearing aid in the directional mode, the listener can focus on the instructor. With the opposite ear hearing aid in the omnidirectional mode, the student will be able to hear others around him or her in the classroom.

Cord, Walden, Surr, and Dittberner (2007) evaluated asymmetric fittings with participants who were fitted with different makes and models of hearing aids. The participants reported greater ease of listening with the asymmetric fitting, but only in those listening situations where they preferred the directional microphone, such as when the signal was close to the listener and background noise was present. In addition, recognition of speech was significantly better with an asymmetric fit than in the omnidirectional mode. However, when comparing a binaural directional fit to the asymmetric directional fit, no significant difference was found in speech recognition performance. In contrast, Hornsby and Ricketts (2007) found that in a noisy condition, binaural directional microphones provided the most optimal listening situation. This study compared binaural directional fittings, binaural omnidirectional fittings, asymmetric directional fittings with the directional microphone in the right hearing aid, and asymmetric directional fittings with the directional microphone in the left hearing aid. Research on this microphone configuration is still limited.

BLUETOOTH AND WIRELESS TECHNOLOGY

Wireless technology in hearing aids is not a recent advancement. The earliest example of wireless technology used in hearing aids was the introduction of induction coils (more commonly referred to as telecoils or "t-coils") in the late 1930s (Levitt, 2007). The use of a small coil in hearing aids coupled with leakage from the electromagnetic field in a telephone allowed patients to wirelessly pair their hearing aids to telephone handsets so as to decrease feedback and the amount of environmental interference during telephone use (Levitt, 2007). As telecommunication technology progressed, so did the use of wireless communication with hearing aids. Currently, most hearing assistive technology (HAT) devices use wireless analog transmission via induction coils, amplitude-modulated (AM) transmission, frequency-modulated (FM) transmission, or combinations of these to assist patients with hearing impairment in difficult listening situations. The introduction of digital wireless technology such as Bluetooth and newer methods of digital magnetic transmission is expanding hearing aid fitting possibilities and the way patients can connect to external devices. Wireless technology is becoming integrated into hearing aids and other instrumentation used in the profession of audiology in many more shapes and forms.

Many audiologists became familiar with Bluetooth technology via the NOAHLink device, which was introduced in 2003 by the Hearing Instrument Manufacturers' Software Association (HIMSA; Ingrao, 2006). This system is used in the fitting process to wirelessly program hearing aids through a personal computer. This device uses Bluetooth technology and communicates to a Bluetooth receiver/transmitter that is located in the computer or attached to the computer via an external emulator. The NOAHLink system, however, still depends on the use of cables to connect with the patient's hearing aids for programming during the fitting process.

The presence of Bluetooth-compatible devices has become profuse within the past few years. Bluetooth is a short-range, digital FM wireless technology that is designed to accomplish wireless connectivity between personal computers, printers, mobile phones, MP3 players, and various other devices (Bridges, 2006; Yanz, 2007a). It has a transmission range between 30—300 ft, depending on the class of Bluetooth (Bridges, 2006).

In the past few years, Starkey Laboratories introduced ELI (see Figure 1), Phonak introduced the SmartLink system, and other companies introduced other Bluetooth accessories to allow patients greater access to cellular phones (Yanz, 2006). Bluetooth accessories pair with Bluetooth devices to send the audio signal to the patient's hearing aids via direct audio input (DAI), FM transmission, electromagnetic relay, or the internal t-coil (Yanz, 2007a).

Recently, hearing aid manufacturers have developed Bluetooth technology within hearing aids specifically for improving connectivity between patients and technology. These new technologies expand the possibilities beyond simply linking hearing aids to cell phones. The Streamer from Oticon A/S is an example of a Bluetooth-compatible central unit that wirelessly pairs the user's hearing aids to a Bluetooth-enabled phone or personal listening device such as an MP3 player, or directly with a connecter wire to a microphone jack input (Schum & Bruun Hansen, 2007). This central unit then sends the signal to the patient's hearing aids via digital magnetic wireless transmission (Schum & Bruun Hansen, 2007). The Streamer will pair with up to eight Bluetooth-enabled devices and is compatible with the Oticon Epoq hearing aids. Similarly, Phonak AG recently introduced a Bluetooth-compatible unit called iCom (see Figure 2) that will pair with up to seven Bluetooth-enabled devices and send this signal wirelessly via magnetic transmission to the patient's hearing aids. This unit has space for an FM receiver to provide patients with improved speech recognition in background noise with the use of a wireless FM microphone. As promising as Bluetooth seems for future use with hearing aids, at the present time, Bluetooth only has the capability for point-to-point (one-to-one) broadcast as opposed to point-to-multipoint broadcast (Yanz, 2007a). This limits the ability of hearing aid users to use Bluetooth in public places such as auditoriums, places of worship, or schools (Yanz, 2007a).

Figure 1. Illustration of the ELI Bluetooth device connected to a direct auditory input boot, which is then connected at the bottom of the hearing aid. Photo courtesy of Starkey Laboratories, Inc. Used with permission.

Wireless communication can certainly improve a user's experience with his or her hearing aids. This technology provides patients access to external devices with clear and low interference transmission and provides improved speech recognition in noise. However, as beneficial as these devices may be, this technology will not prove helpful if hearing aid patients do not take advantage of its availability and what it has to offer. Many hearing aid users are reluctant to purchase additional accessories, and the use of external devices, even today, is not appealing to all hearing aid users. External devices are required to access Bluetooth technology due to certain hearing aid limitations. Currently, it is not possible to house a Bluetooth chip within the hearing aid due to chip size and battery power drain requirements.

Over the years, improvements in hearing aids and hearing aid features have centered on the same goal: improving user satisfaction with hearing aids by improving comfort and speech recognition in noise. Advances in wireless communication and hearing aid technology have allowed manufacturers to design hearing aids that communicate between each other (i.e., between left- and right-ear instruments). Several hearing aid manufacturers have recently introduced products incorporating wireless communication between hearing aids and, undoubtedly, many more will do so in the future. For example, Oticon A/S recently introduced the Epoq hearing system with a form of digital magnetic wireless communication at the chip level that allows the hearing aids in a binaural fitting the ability to coordinate settings such as switch position, microphone mode, and program selection (Schum & Bruun Hansen, 2007). Depending on the speed of this transmission, the hearing aids have the ability to relay discrete pitch and amplitude cues between one another to assist with improvements in binaural hearing and localization (Schum & Bruun Hansen, 2007). This effort to preserve binaural cues is an attempt to improve speech recognition in noise.

Figure 2. Illustration of the iCom Bluetooth device that was recently introduced by Phonak. This device hangs around the neck of the patient and allows for wireless communication with the hearing aids. Photo courtesy of Phonak, Inc. Used with permission.

Currently, Bluetooth technology is too large to be incorporated in hearing aid designs. As magnetic or Bluetooth digital wireless capabilities advance, the application to hearing aids will also progress. Processing speed and increasing data rate will facilitate faster ear-to-ear transmission. Advances in decreasing the chip size and power consumption of Bluetooth technology may allow for future Bluetooth receivers to be embedded in all hearing aids (Edwards, 2007). Also, Bluetooth may one day become standard in public systems such as airports and theaters, providing countless improvements for patients with hearing impairment in these listening environments.

NOISE REDUCTION

Noise reduction algorithms have undergone significant changes since the inception of this technology. Generally, early analog noise reduction algorithms were intended to filter out noise by reducing low-frequency gain. This was an effort to prevent strong low-frequency signals from activating the hearing aid's compression circuitry and potentially reducing gain for all frequency regions (Bentler & Chiou, 2006). The limitations of analog noise reduction algorithms coincided with the limitations of analog circuitry—noise reduction schemes were implemented in one channel and the gain reduction was often based on input level only (Bentler & Chiou, 2006).

The introduction of DSP allowed for the use of more complex algorithms for noise reduction. The earlier strategies for digital noise reduction (DNR) were the Wiener filter (Wiener, 1949) and spectral subtraction (Anzalone, Calandruccio, Doherty, & Laurel, 2006; Bentler & Chiou, 2006). The Wiener filter estimates the characteristics of speech and noise and creates a filter that optimizes the SNR at the output filter (Anzalone et al., 2006). In spectral subtraction, the noise spectrum is obtained during pauses in speech, and this is subtracted from the speech-plus-noise spectrum, leaving only the spectrum of speech (Anzalone et al., 2006; Bentler & Chiou, 2006). The limitations of these schemes were that stationary speech and noise signals were needed for the Wiener filter to operate, and audible distortions were needed for spectral subtraction to work (Bentler & Chiou, 2006).

Unlike with directional microphones, DNR algorithms rely on differences between the physical characteristics of speech and noise rather than spatial separation (Chung, 2004). Speech has a modulation rate that occurs roughly between 3 Hz and 10 Hz (Bentler & Chiou, 2006; Chung, 2004; Schum, 2003), with most fluctuations occurring around 3 Hz (Bentler & Chiou, 2006). Often, noise will be more stable or will modulate at a different rate. Speech modulation also varies in amplitude over time. Gain reduction is applied, but DSP allows the modulation of speech to be monitored in various channels, and as a result, gain reduction can be more specific.

DNR algorithms do not separate speech from noise. Rather, DNR simply determines whether or not a channel is dominated by noise (Schum, 2003). Gain is reduced in each frequency channel based on the modulation or lack of modulation of the unwanted noise signal and that of the desired speech signal. The decision to make these reductions in gain are based not solely on the modulation differences between speech and noise, but on other factors including the SNR in each channel, the weighting of specific frequencies with regard to speech information, and the onset and offset times of the DNR algorithm (Chung, 2004). If the SNR is set too high for speech-in-noise situations, then there is a risk of allowing too much noise to pass through. If the SNR is set too low, then there is a risk of reducing audibility (Bentler & Chiou, 2006). Furthermore, if the SNR is high, the speech signal strength is higher than the noise, and the calculations will result in no noise reduction in that channel (Chung, 2004).

Generally, the mid-frequencies are considered to hold the highest contribution with regard to speech recognition. Some DNR algorithms may apply this theory and choose to have areas of less gain reduction or less compression in selected frequency channels in an attempt to preserve speech information. The timing, or onset and offset, of DNR algorithms is also an important consideration. Few studies have yet been conducted to investigate this, but it may be theorized that too slow of a response time could cause the algorithm not to respond to sudden noises in the environment, and too fast of a response time could cause fast speech sounds to be reduced in audibility (Bentler & Chiou, 2006; Chung, 2004).

Multichannel noise reduction algorithms are based on modulations in each frequency channel to determine the presence of speech. If the differences between acoustic characteristics of the speech and competing signal are narrow, such as when speech is the competing noise, the DNR may not be as effective as when the noise is steady state (Chung, 2004). The multichannel approach to noise reduction certainly provides advantages. However, the more narrow the filters, the greater the possibility of group delay (Nordrum, Erler, Garstecki, & Dhar, 2006). Group delay is the amount of processing time that a circuit takes to transport a signal from the microphone to the receiver. If the delay is too long, there could be potential negative effects, such as a perceived echo by the listener (Nordrum et al., 2006). The rules applied to DNR are unique to each manufacturer. Manufacturers will vary in the SNR used, weighting the importance of the speech frequencies, timing strategies, and even classification of noise. How noise is classified will affect all properties of DNR for that hearing aid. It is important for audiologists to understand the basic rules of DNR and how manufacturers use and apply these rules.

Another algorithm employed for DNR is synchrony detection. Synchrony detection is based on the comodula-tion of speech when the vocal folds open and close. The fundamental frequency of the human voice is created by the vibration of the vocal cords, up to approximately 250 times per second, or 250 Hz, for adults. Vowels and voiced consonants are composed of harmonics of the fundamental frequency, which essentially are the bursts of periodic and synchronous energy that result when the vocal folds open and close (Chung, 2004; Schum, 2003). Synchrony detection, which is currently used by only one manufacturer, bases the decision of gain reduction on the lack of presence of this synchronous energy. In other words, this gain reduction strategy is based on the absence of speech. This system tracks the synchronous energy across specified high-frequency channels and provides amplification, or decreases compression, when speech is present and increases compression or reduces gain when speech is not detected (Schum, 2003). This algorithm also has limitations with regard to improving speech recognition when the signal and noise are similar in their spectral patterns (i.e., when the competing signal is speech; Chung, 2004). This particular manufacturer has introduced products that combine the use of the multichannel DNR and synchrony detection algorithms.

Regardless of the algorithm applied, the main goal of DNR is to improve speech recognition in background noise and improve ease of listening in noise (Kuk, Peters, Keenan, & Lau, 2007; Mueller, Weber, & Hornsby, 2006; Schum, 2003). Most hearing aids contain a DNR algorithm; use of this feature has become fairly standard across manufacturers. Most research to date, however, is inconsistent with regard to improvements in speech-in-noise performance with the use of DNR (Bentler & Chiou, 2006). Also, most studies have not isolated DNR when evaluating the use of recent hearing aid features and speech recognition in noise. For example, many reports on the effectiveness of DNR have actually reported on the effectiveness of DNR combined with directional microphones. The combined use of DNR and directional microphones is likely how most patients use their hearing aids. Nordrum et al. (2006) performed a study to compare performance on the Hearing in Noise Test (HINT; Nilson, Soli, & Sullivan, 1994) with a directional microphone combined with DNR, and with each feature isolated. Their findings were similar to past reports: There was not a noticeable improvement in performance of speech-in-noise when using DNR in isolation.

A relatively new measure used to asses a listener's reaction to background noise is the acceptable noise level (ANL; Nabelek, Tuckler, & Letowski, 1991). The ANL is the difference between the most comfortable listening level for speech and the highest acceptable level of background noise when listening to a story (Nabelek, Tampas, & Burchfield, 2004). In general, if a person has good acceptance of background noise, then his or her measurement will reveal a smaller ANL. People with greater ANLs will tend to have lower acceptance of background noise (Nabelek et al., 2004). Mueller et al. (2006) used the HINT to assess the effects of DNR, in isolation, on a person's ANL and speech intelligibility. The results, under specific test conditions, revealed a clinically significant improvement in ANL scores with the DNR turned on versus the DNR turned off. Assessment of speech intelligibility revealed results similar to previous studies; no clinically significant improvement in speech intelligibility was measured with the use of DNR. Speech intelligibility was not affected positively or negatively with DNR activated or deactivated.

Although the ultimate goal of DNR is to improve speech recognition in noise, research has generally not shown this to be successful. Currently, DNR is best at providing comfort in noise. As DSP chips improve and processing speed increases, newer and more complex DNR algorithms may be applied. Until then, it is important for audiologists to understand how DNR differs across manufacturers and how various algorithms may affect patients.

SINGLE-SIDED DEAFNESS (SSD)

Patients with SSD typically have had limited options with regard to amplification. In the past, these patients were offered communication strategies such as sitting with the good ear closest to the speaker and reducing background noise whenever possible and/or were recommended to consider wireless or wired analog contralateral routing of signal (CROS) hearing aids. SSD is typically characterized by a severe or profound sensorineural hearing loss, poor word recognition score, and/or significantly decreased dynamic range resulting in intolerance for loud or amplified sound in one ear and normal hearing sensitivity in the opposite ear (Valente, Valente, & Mispagel, 2006). The major communication difficulties reported by patients with SSD are recognizing speech in background noise, localization, and recognizing speech arriving from the impaired side (Valente, Valente, & Mispagel, 2006; Wazen, Ghossaini, Spitzer, & Kuller, 2005).

The introduction of multichannel DSP, improved feedback cancellation, improved wireless technology, and new bone conduction devices have expanded the possible amplification options for these patients. All of the available systems use the hearing in the better ear. In this fitting strategy, the cochlea of the better ear receives the signal from the poorer ear via air conduction, by wired or wireless CROS device, or via bone conduction using a bone conduction hearing aid, transcranial hearing aid, TransEar device, or bone-anchored hearing aid (BAHA).

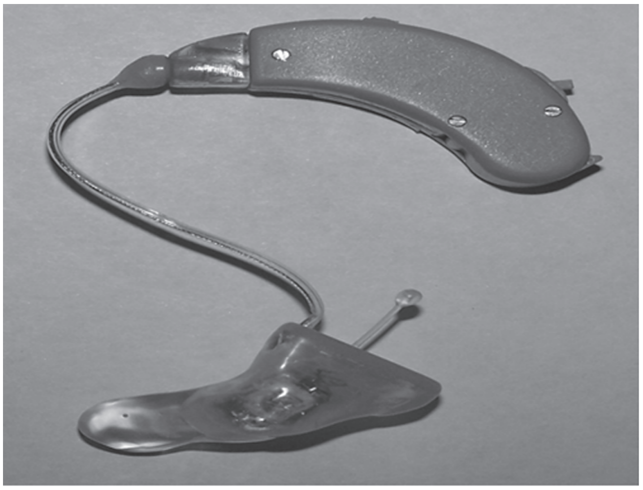
Telex Inc. was the first company to introduce a commercially available wireless CROS hearing aid in 1973 (Valente, Valente, Enrietto, & Layton, 1996). At that time, hearing aid technology was analog but offered freedom from being connected via a wire. These devices, typically in-the-ear (ITE) to ITE or behind-the-ear (BTE) to BTE configurations, have a transmitter microphone positioned on the side of the poorer ear that sends sound arriving at that side to a receiver located at the better ear, typically via AM radio waves. Analog technology limited the ability of the audiologist to shape the frequency response of the signal that was received by the amplified or better ear (Hayes, Pumford, & Dorscher, 2005). Wireless CROS devices are now available with multichannel and multiband DSP, allowing the possibility of greater flexibility in the adjustment of the frequency response (Hayes, Pumford, & Dorscher, 2005). Other features available in these hearing aids include automatic adaptive directional microphones, noise reduction, datalogging, and multiple memories.

Phonak, Unitron, and Interton offer wireless CROS/BiCROS options. The Phonak and Unitron CROS/BiCROS hearing aids use a 374-kHz AM signal, and the Interton system uses a 1.8-MHz AM signal. The AM signal is analog; thus, the transmission is prone to interference from strong electrical signals or other AM signals. Anecdotally, some patients have reported walking by an AM radio tower and clearly hearing signals being broadcasted by the station. The strength of the transmission can also be affected by the distance between the transmitter and the receiver. A smaller head circumference may yield a stronger signal (Hayes et al., 2005). Another limitation of CROS devices is related to the hearing in the better ear. According to Valente, Potts, Valente, and Goebel (1995), patients had a better acceptance with CROS amplification if slight hearing loss was present above 1500 Hz. In other words, patients with normal hearing in the better ear may be more inclined to reject CROS amplification.

A transcranial CROS fitting takes advantage of interaural attenuation (IA). IA is the least amount of signal intensity needed to cross through the temporal bone via bone conduction, as well as around the head via air conduction from an ear with a moderately severe or worse hearing loss to the cochlea of a significantly better hearing ear (Valente, Valente, & Mispagel, 2006). A power hearing aid is fit to the ear with poorer hearing, and the amplified signal is intended to be strong enough to pass through the skull and reach the cochlea of the better hearing ear via air and bone conduction. Valente, Potts, et al. (1995) reported that participants who returned the transcranial CROS complained of feedback and vibration. Current DSP hearing aids with improved feedback cancellation and new shell technology may assist with eliminating feedback that prevented success with the previous versions of transcranial CROS using analog signal processing. Transcranial CROS may be acceptable to many patients because only one device is required. Also, patients with normal hearing thresholds in the opposite ear seem to perform better with this approach (Valente, Valente, & Mispagel, 2006). Valente, Valente, and Mispagel (2006) called this a quasi-transcranial CROS fitting because the amplified signal actually reaches the cochlea by using air conduction signals at loud intensity levels to vibrate the skull and by taking advantage of IA to send the signal to the normal hearing cochlea.

True transcranial CROS is obtained through bone conduction. The IA for bone conduction is 0 dB; therefore, very little gain/power is needed for the better ear to receive the amplified signal from the poor ear. Traditional bone conduction aids are still available in a BTE, eyeglass, or body-worn device, with the oscillator held firmly against the mastoid with a headband or as an eyeglass. A newer and less cumbersome version of a bone conduction aid is the TransEar by Ear Technology Corporation (see Figure 3). The TransEar consists of a digital BTE hearing aid, a connector wire, and an acrylic earmold or "transfer unit" that houses a miniature bone conduction oscillator (TransEar, 2008). The TransEar is placed in the ear canal of the poorer ear. The earmold must have a long bore and is much tighter than an earmold used in a conventional air conduction fitting due to the pressure needed to transfer the signal from the oscillator through the earmold to the bones of the skull. In addition, a tight seal for the earmold in the ear canal is needed in order to obtain sufficient gain without feedback or vibration. Anecdotally, patients not having a proper seal have reported experiencing feedback or vibration from the oscillator when louder input levels are present. Bone conduction hearing aids and the TransEar have to overcome the interference presented by the tissue between the oscillator and the bones of the skull before the signal reaches the cochlea. These devices provide a non-invasive, true transcranial CROS option.

Figure 3. Illustration of the TransEar from Ear Technology Corporation. The digital signal processing hearing aid (upper) connects to the custom earmold (lower) via a wired link (available in seven lengths). In the earmold is a bone conduction receiver that transmits sound from the ear canal of the "dead" ear across the skull to the cochlea of the normal inner ear on the opposite side. Photo courtesy of Ear Technology Corporation. Used with permission.



The Baha is an osseointegrated auditory prosthesis that was approved by the Food and Drug Administration (FDA) in 2002 for use by patients with SSD and is available through Cochlear Americas (Valente, Valente, & Mispagel, 2006). Patients undergo outpatient surgery under local or general anesthesia to have a small titanium screw and abutment placed in the mastoid process of the poorer hearing ear. This implant must become fully integrated with the bone before the processor can be attached (see Figure 4). In the adult patient, this takes 3 months; in the pediatric patient, it may take as long as 6 months. Children cannot be implanted until the age of 5 because the skull is still too soft for the abutment to stay in place. Until then, children can use a soft band with the processor. The abutment receives the vibrations from the processor and transfers the amplified signal to the implant, which then sends the amplified signal directly to the cochlea of the opposite ear (Cochlear Americas, 2007). Because the implant becomes part of the mastoid bone, there is no skin or tissue to impede the strength of the signal. The processor attaches to the abutment, and there is no physical pressure exerted against the head or in the ear canal as might be experienced with a traditional bone conduction aid or the TransEar. The Baha is available in two head-worn DSP devices, the Divino and the Intenso, and an analog body-worn processor, the

Figure 4. Illustration of the Baha (right) along with the titanium screw and abutment (left) that the Baha snaps into. Photo courtesy of Cochlear Americas. Used with permission.

Cordelle II (Cochlear Americas, 2007). All processors have electrical inputs to allow connection to personal FM systems, personal listening devices, and some mobile phones. When patients meet candidacy criteria for the Baha, this is an excellent alternative for patients with SSD.

No single device is appropriate for all patients with SSD; however, the advances in technology and fitting strategies now allow audiologists greater options for patients to consider.

FREQUENCY TRANSPOSITION

Frequency transposition is an option that was created for individuals with severe to profound high-frequency hearing loss who are likely not to receive any benefit from high-frequency amplification. These individuals have significant difficulty with recognizing speech. The theory behind frequency transposition is to transform the amplified signal in the high-frequency region and shift this segment to the lower frequencies where hearing is better in order to make speech more audible.

Hearing aids designed with frequency transposition have been reported to benefit some patients. In a study performed by Parent, Chmiel, and Jerger (1998), the TranSonic Device improved speech recognition for 2 of the 4 participants. However, the device was large and cumbersome to the participants. McDermott, Dorkos, Dean, and Ching (1999) stated that the improvements reported by the participants using the AVR TranSonic frequency-transposing hearing aid were mostly from the low-frequency electroacoustic characteristics of the hearing aid. The AVR ImpaCT was the first BTE frequency transposition aid but did not prove to be beneficial in improving high-frequency speech information (McDermott & Knight, 2001).

Widex has taken frequency transposition and added this feature, referred to as audibility extender, to their Inteo line of hearing aids. The fitting software selects a start frequency where the transposition begins based on the patient's sensogram. The sensogram is the in-situ audiogram performed within the Widex software. A linear transposition algorithm is used to transpose sounds by one octave below the start frequency into audible frequency regions. The sounds are moved from high-frequency regions to middle- and low-frequency regions, thus allowing the user to perceive higher frequency consonants and sounds that may have previously been inaudible (Kuk et al., 2007). By having the audibility extender as a program option, the user can choose to use the audibility extender as the primary listening mode or as a secondary program for situations in which he or she needs to be aware of important high-frequency sounds.

Phonak has introduced frequency compression in its Naida hearing aid model. This hearing aid uses a cutoff frequency as a point to stop amplifying and begin compressing high-frequency sounds into the lower frequency regions. This hearing aid, however, is currently designed only for those individuals with severe to profound hearing loss. Peer-reviewed research has not yet been published regarding the effectiveness of this feature.

SELF-LEARNING AND ADAPTATION/ACCEPTANCE MANAGERS

Like cell phones, some hearing aids now incorporate self-learning features. For example, the hearing aid monitors when the user adjusts the volume control and after a certain period of time, the hearing aid automatically adjusts itself to the average level where the user typically sets the volume control (Hayes, 2007). Some hearing aids will also remember where the patient had his or her volume set and increase the volume by a small number of decibels until it eventually reaches a preset target that was established at the time of the hearing aid fitting.

This feature lets the patient have control of the hearing aid while at the same time allows the patient to gradually adapt to the necessary gain levels. Usually, a patient's hearing aids are fit using real ear measurements (REM) and initially, the patient may report that the hearing aids are too loud. The gain settings are subsequently reduced to adjust for comfort. Over time, the patient usually adjusts to the volume settings and perceives the hearing aids as not being loud enough. As a result, additional programming by the audiologist is needed, requiring additional visits to the clinic.

In earlier years, some hearing aid manufacturers tried to combat this issue by having a step-based system in their fitting software that was based on the patient's experience level. As the patient acclimatized to the hearing aid, the hearing aid was gradually moved up on the experience ladder to increase gain, again requiring additional visits to the audiologist.

For those patients who have difficulty getting to the audiologist's office or who live far away, these extra visits can be a burden. With the use of the automatic self-learning in the hearing aids, the gain can gradually be increased as the patient acclimatizes to the hearing aids without having to return to the office. Hearing aids in the future may provide a true intelligent form of automation that will respond to the particular environment of the individual. In the future, this technology will not only adjust the volume setting, but will take into consideration the listening environment and various listening situations the patient frequents. The hearing aid will make adjustments to the volume and frequency settings based on information that it receives (Edwards, 2007).

INTEGRATED REM

For decades, REM have been used for verifying hearing aid fittings by placing a probe microphone in the patient's ear and recording the acoustic responses with and without the hearing aid in place. Because of the need to obtain accurate, objective information during the hearing aid fitting, the American Academy of Audiology has included REM as an important component of the Best Practice Standard (Valente, Abrams, et al., 2006).

Despite the importance of using REM, most hearing health care professionals have been reluctant to routinely incorporate REM into their hearing aid fittings. A survey conducted in 1998 stated that although 75% of audiologists own probe microphone equipment, only 42% to 45% of these practitioners report actually using it (Mueller, 1998). Many practitioners find the equipment required to perform REM expensive and too large for their clinics. Some professionals also find the wires and extra equipment too cumbersome and time consuming during a hearing aid fitting. Finally, some hearing care professionals may find REM to be redundant and unnecessary because additional adjustments are usually made to the hearing aids based on patient report of clarity, loudness, comfort, and sound quality, after the REM are obtained (Yanz, 2007b).

Recently, one hearing aid manufacturer, Starkey Laboratories, integrated REM into the circuitry of two of its hearing aid models. The hearing aid uses the programming software to assist in performing the REM. This development takes away the need for extra equipment in the office, saving space and money. It also is less cumbersome because all the pieces required to conduct REM are incorporated into the hearing aid. The receiver provides a calibrated signal and the microphone serves as the probe microphone. A thin probe tube is placed on the microphone of the hearing aid and the other end is placed in the patient's ear. The hearing aid produces the signal to measure the acoustic responses, and the software guides the audiologist through the fitting process. The measurements take less than a minute and reduce the amount of time spent on REM (Yanz, 2007b).

This is a significant development in helping to make the standard of practice easier, more affordable, and more convenient to hearing care professionals. However, there are a few limitations. At the present time, only one manufacturer (Starkey) has made this feature available, and it is limited to the Destiny 1600 and Zon 7, which are high-end models that may not be affordable to some patients. Also, peer-reviewed research is not yet available to confirm whether integrated REM results are comparable to those obtained in the conventional manner.

CONCLUSION

There have been significant improvements in hearing aids over the past few years. These improvements have allowed audiologists to provide amplification to patients who previously may not have been able to use hearing aids. Recent improvements also allow for greater ease and flexibility in the fitting process. Directional microphones and noise reduction algorithms have improved over the past several years, and wireless technology has been reintroduced into hearing aids. With the progression of microelectronics, the future of hearing aids looks quite promising. As the processing speed and memory in hearing aids increase, the algorithms will become more sophisticated (Edwards, 2007). Speech recognition may improve, and amplified sounds may be perceived as being more natural (Launer, 2008). In addition, the connectivity with other audio sources will become more sophisticated and widespread (Launer, 2008).

Future developments in hearing aids will become more complex and will require combined efforts of psychoacoustics, signal processing, and clinical research conducted by audiologists (Edwards, 2007). These improvements will attempt to better mirror the complex nature of hearing loss and use patient feedback to help achieve goals of providing improved devices for the population with hearing impairment (Edwards, 2007).

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