Challenges and Recent Developments in Hearing Aids

Part II. Feedback and Occlusion Effect Reduction Strategies, Laser Shell Manufacturing Processes, and Other Signal Processing Technologies

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This is the second part of a review on the challenges and recent developments in hearing aids. Feedback and the occlusion effect pose great challenges in hearing aid design and usage. Yet, conventional solutions to feedback and the occlusion effect often create a dilemma: the solution to one often leads to the other. This review discusses the advanced signal processing strategies to reduce feedback and some new approaches to reduce the occlusion effect. Specifically, the causes of three types of feedback (acoustic, mechanical, and electromagnetic) are discussed. The strategies currently used to reduce acoustic feedback (i.e., adaptive feedback reduction algorithms using adaptive gain reduction, notch filtering, and phase cancellation strategies) and the design of new receivers that are built to reduce mechanical and electromagnetic feedback are explained. In addition, various new strategies (i.e., redesigned sound delivery devices and receiver-in-the-ear-canal hearing aid configuration) to reduce the occlusion effect are reviewed. Many manufacturers have recently adopted laser shell-manufacturing technologies to overcome problems associated with manufacturing custom hearing aid shells. The mechanisms of selected laser sintering and stereo lithographic apparatus and the properties of custom shells produced by these two processes are reviewed. Further, various new developments in hearing aid transducers, telecoils, channel-free amplification, open-platform programming options, rechargeable hearing aids, ear-level frequency modulated (FM) receivers, wireless Bluetooth FM systems, and wireless programming options are briefly explained and discussed. Finally, the applications of advanced hearing aid technologies to enhance other devices such as cochlear implants, hearing protectors, and cellular phones are discussed.

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1. Introduction

Part I of the review (Chung, 2004) discussed the challenges of listening to speech in noise and recently developed algorithms and hardware to increase speech understanding and listening comfort in noise. This part of the review discusses the challenges and newly developed strategies and algorithms to reduce feedback and the occlusion effect. Feedback is commonly referred to as whistling sounds emitted by hearing aids. The occlusion effect is caused by the trapping of bone-conducted energy in the ear canal when the ear canal is occluded. The most common complaints of the occlusion effect are perceived unnaturalness of the user’s own voice (e.g., talking in a barrel) and unusually loud chewing sounds.

Although feedback and the occlusion effect appear to be two unrelated events, the common conventional strategies used to reduce feedback (most often by decreasing the vent size) and the occlusion effect (most often by increasing the vent size) impose a dilemma in the hearing aid fitting process. Fortunately, as hearing aid signal processing technologies become more sophisticated, advanced feedback reduction algorithms are developed to cancel or reduce feedback, thus allowing a larger vent to reduce the occlusion effect. These advanced feedback reduction algorithms have also brought about newly designed open ear canal-fitting sound delivery devices that are aimed at further reducing the occlusion effect.

This review discusses the types of feedback, causes, conventional management strategies, and newly developed feedback reduction algorithms, such as adaptive gain reduction, notch filtering and phase cancellation, and their limitations. In addition, redesigned receivers with pantograph, extra padding, and a metal can to reduce mechanical and electromagnetic feedback are explained. Further, new sound-delivering devices for open ear-canal fittings and deep canal fittings to reduce the occlusion effects are discussed. Verification data, if available, are also summarized.

Recent advances in three-dimensional (3D) laser scanning and printing technologies have revolutionized the manufacturing and prototyping processes. Parts from military jet components, Formula 1 car parts, medical implants and prosthesis, to full-colored geographic models for military planning purposes can be made in laboratories within several hours (Amato, 2003). These instant manufacturing and rapid prototyping technologies have also made a big impact on the custom hearing aid shell manufacturing process. According to hearing aid manufacturers, these technologies have greatly reduced the time and the manpower required in the shell manufacturing process and have improved the accuracy of custom shell fittings. This review describes the mechanisms of two laser printing technologies, namely, selective laser sintering and stereo lithographic apparatus. Also discussed are validation data and the limitations of the custom shells manufactured by these two processes.

In addition to the above three major areas, various new developments in hearing aid hardware (e.g., microphones, receivers, telecoils, rechargeable hearing aids), signal processing strategies (e.g., channel-free compression, open-platform hearing aids, and binaural hearing aids), programming technologies (e.g., Bluetooth wireless programming devices), and personal frequency modulated (FM) systems (e.g., behind-the-ear FM receivers, versatile FM systems with Bluetooth cellular phone compatibility) are discussed. Last, new applications of advanced hearing aid technologies to enhance cochlear implants, hearing protectors, and cellular phones are also briefly mentioned.

2. Challenge No. 2: Feedback

Feedback is one of the most frequent complaints of hearing aid users (Kochkin, 1994). The three distinctive types of feedback in hearing aids are acoustic, mechanical, and electromagnetic. The most common type is acoustic feedback, which is often referred to as high-frequency tonal sounds emitted by hearing aids. In a more general sense, it is the output of the receiver escaping from the ear canal, tubing wall, and the sides of earmold/shell/vent being picked up by the hearing aid microphone, whether or not a whistling sound is heard. A whistling or ringing feedback signal is heard only when the hearing aid goes into “oscillation.” This happens when the amplification from the microphone to the receiver is greater than the attenuation of the amplified sounds from the receiver to the microphone (Hellgren et al., 1999a; Lybarger, 1975). The frequency range of acoustic feedback is usually 2 to 5 kHz (Dyrlund and Lundh, 1990; Langford-Smith, 1960).
Mechanical feedback occurs when the vibration of the receiver is transmitted to the microphone via the shell and faceplate of an in-the-ear hearing aid or when the vibration of tubing is transmitted to the microphone of a behind-the-ear hearing aid (Agnew, 1996; Thompson, 2002). The faceplate/tubing then acts as a vibrating surface that vibrates the surrounding air molecules. Mechanical feedback occurs if the vibration is picked up by the microphone, is reamplified in the signal processing path, and forms a feedback loop.

Electromagnetic feedback takes place when the magnetic radiation emitted by the receiver is picked up by the telecoil and then reamplified (Agnew, 1996; Thompson, 2002). This type of feedback occurs when the hearing aid is in the telecoil mode.

Different signal processing algorithms and strategies have been developed in recent years to reduce feedback in hearing aids. The next section discusses the causes of the three types of feedback in more detail and explains various algorithms and strategies used to reduce feedback.

2.1. Acoustic Feedback

Acoustic feedback poses multiple challenges to clinicians and hearing aid users:

1. Feedback limits the amount of maximum gain that can be provided by a hearing aid and limits the style of the hearing aid that can be fitted to users with a particular degree of hearing loss.
2. Feedback limits the size of the vent on a hearing aid and may generate negative effects such as the occlusion effect or problems caused by a poorly ventilated ear canal.
3. When the gain of a hearing aid is set close to just below the gain level that generates feedback, sub-oscillatory feedback is generated. The consequences include peaks and valleys created in the frequency response of the hearing aid, high-frequency whistles or overtones present in the processed speech, and degraded speech intelligibility and sound quality (Cox, 1982; Engebretson and French-St. George, 1993).
4. Whistling feedback not only affects speech intelligibility and perceived sound quality but also can be very annoying and embarrassing to hearing aid users because they and/or people around them can hear the high-frequency tones. This can be especially embarrassing for those cosmetically conscious users who want to conceal their hearing loss. If the feedback problem is not resolved, it often leads to in-the-drawer hearing aids.

2.1.1. Causes of Acoustic Feedback

Acoustic feedback can arise from various different sources:

1. user characteristics, such as the amount of ear canal resonance, the shape and the size of the pinna, and mandibular movements;
2. physical characteristics of hearing aids, such as the style of the hearing aid, venting and tubing choices, and the canal length of the earmold/shell;
3. changes in the acoustic environment, such as talking, chewing, yawning, receiving a hug, placing a telephone handset close to the hearing aid; and
4. hearing aid or earmold malfunctions and defects, such as microphone tube leaks, internal components contacting each other, receiver tube leaks, internal vent cracks and holes, and a crack in the earmold tubing. (For a detailed review on the causes of feedback, please see Agnew, 1996.)

In addition, multichannel hearing aids with wide dynamic range compression (i.e., programmable and digital hearing aids) are generally more prone to feedback than linear hearing aids or hearing aids with compression limitors (Kuk et al., 2002; Olsen et al., 2001). This is because hearing aids with wide dynamic range compression generally provide more gain for low-level sounds and less gain for high-level sounds. Thus, feedback may be generated when the user is in a quiet environment or when the level of the incoming signal is low. Feedback may subside when the level of the incoming signal increases as the hearing aid provides less gain for higher level sounds (Figure 1).

2.1.2. Conventional Strategies to Reduce Acoustic Feedback

In the long history of hearing aid development, many strategies have been used to reduce acoustic feedback. To resolve feedback due to hearing aid or earmold malfunctions and defects, the hearing aid is usually sent back to the manufacturer for repair or the earmold and the tubing...
are modified or re-serviced in office. The management of feedback caused by the physical characteristics of hearing aid and changes in acoustic environment is complicated by the occlusion effect, which is caused by trapped bone-conducted sounds in the ear canal (Killion et al., 1988; von Bekesy, 1960; Zwislocki, 1953). In theory, acoustic feedback can be effectively reduced by reducing the vent size or plugging the vent if the occlusion effect does not exist. In practice, either strategy significantly increases the occlusion effect, which creates a booming sensation (e.g., talking in a barrel) when the hearing aid user speaks, chews, swallows or even walks.

Because of the dilemma between feedback and the occlusion effect, traditional feedback management approaches usually reduce the gain of the forward path and increase the attenuation of the feedback path. Methods used to reduce the gain of the forward path include turning the volume control down, reducing the gain at one or more high-frequency bands in which feedback occurs, reducing the gain for low-level inputs of a wide dynamic range compression hearing aid, and changing to binaural fitting to reduce the gain needed for each ear. Methods used to increase the attenuation of the feedback path include reducing the vent size (to a certain extent), increasing receiver tube damping, tightening the earmold or lengthening the canal part of the hearing aid, and changing the hearing aid style.

Although feedback can be effectively resolved in some cases, many of these traditional feedback management strategies have associated problems:

1. Reducing the vent size often results in an increase in the occlusion effect, as previously mentioned.
2. Tightening the earmold or lengthening the canal portion of the hearing aid often leads to discomfort in hearing aid use (Kuk and Ludvigsen, 2002b; Bongiovanni et al., 1991; Pirzanski, 1998).
3. Changing the hearing aid style may not be acceptable to the user and results in infrequent use of the hearing aids or “in-the-drawer” hearing aids.

Figure 1. Feedback versus level in a wide dynamic range compression hearing aid. Feedback is present at the lowest input level but disappears at higher input levels.
4. Lowering the volume control of the hearing aid or permanently reducing the gain in one or more frequency bands or channels may result in under-amplification of speech signals.

If lowering the gain is inevitable, a hearing aid with more channels or bands generally has less negative effects because feedback reduction can be achieved by reducing the gain in a relatively narrower frequency region (Mueller, 2002). The more the bands or channels, the finer the control and the less the overall signal reduction. Yet, digital hearing aids with many channels may have longer processing delays as discussed in Part I of this review.

In addition to these strategies, researchers and engineers have developed many electronic solutions in an attempt to reduce acoustic feedback, for example, electronic damping of high-frequency peaks, bandpass filtering, notch filtering, frequency shifting, frequency warbling, and the adaptive cancellation filter. For brief discussions on their mechanisms and effectiveness, please refer to Agnew (1996).

As technologies advance, multiple adaptive feedback reduction algorithms have been developed to reduce feedback using different strategies. The common goal of these adaptive feedback reduction algorithms is to reduce feedback without permanently affecting the gain of hearing aids. The next section discusses some of these strategies.


The new generation of adaptive feedback reduction algorithms uses digital signal processing to reduce feedback. Some adaptive feedback reduction algorithms act on the feedback signal when the hearing aid goes into oscillation and generates a whistling sound. Others act on the transfer function (frequency-amplitude-phase relationship) of feedback signals even before the hearing aid generates a whistling feedback signal. An example of a frequency response (frequency-amplitude relationship) of a silent feedback signal is shown in Figure 2. Notice that the magnitude of the feedback signal is less than 0 dB gain level.

![Figure 2](image-url)  
*Figure 2.* The frequency response of the feedback path of a vented behind-the-ear hearing aid when a telephone handset is placed near the ear (solid line) and removed (dashed line). Reprinted with permission from Kates (2003), Journal of Acoustical Society of America, 106.
and the frequency response is not peaky (i.e., no feedback loop is formed).

To distinguish the two types of acoustic feedback, the next section uses whistling feedback signal to refer to the type of feedback that is audible to either the hearing aid user or people in the surroundings. The type of feedback that is generated without an audible whistling signal is called an inaudible feedback signal. When both types are referred to, the term feedback is used.

2.1.3.1. How They Work

As mentioned, feedback can be caused by hearing aid or user characteristics (static component) or by abrupt changes in the acoustic environment (dynamic component). A common implementation of the new generation of adaptive feedback reduction algorithms has a fixed/slow-acting and an adaptive/fast-acting component to act on the static and dynamic components of feedback, respectively.

The fixed/slow-acting component increases the algorithm’s stability by reducing feedback arising from the static component. On the other hand, the adaptive/fast-acting component enhances the effectiveness of the feedback reduction algorithm by reducing feedback that arises from abrupt changes in the acoustic environment. Depending on individual differences and how close the reflective surface is, these changes can alter the magnitude of feedback by 5 to 10 dB and sometimes by 20 dB or more (Rafaely et al., 2000; Hellgren et al., 1999b).

Although various feedback reduction signal processing approaches have been explored in the past, the three primary adaptive approaches that have been implemented in current commercially available hearing aids are adaptive gain reduction, adaptive notch filters, and phase cancellation. The mechanisms of these feedback reduction algorithms are summarized in Table 1.

2.1.3.1. Adaptive Feedback Reduction Algorithm by Gain Reduction

Algorithms that reduce feedback by reducing the gain of the hearing aid are often referred to as feedback management algorithms/systems. The general approach is to reduce the gain of the frequency channel(s) in which feedback occurs. The difference in newer adaptive feedback management systems is that their gain reduction is adaptive, instead of fixed as in the conventional systems. Adaptive systems reduce the gain of a frequency channel only if feedback is detected in that channel, and the amount of gain reduction is variable, depending on the magnitude of the feedback signal. In addition, newer systems use different strategies to reduce the gain for low-level inputs only, rather than for all input levels as in conventional systems. Two examples of these strategies are shown in Figure 3.

A specific example of an adaptive feedback management system is found in Phonak Perseo and Valeo hearing aids, which have an optional feedback manager test and an adaptive gain reduction component. Clinicians can conduct the optional feedback manager test in the initial hearing aid fitting session. During the test, the gain of a channel is automatically set to the maximum, and the feedback detector monitors the presence or absence of tonal (e.g., pure tone) and periodic signals. If a tonal signal is not detected in a frequency channel, the maximum gain of the channel at low level is limited by the particular model of the hearing aid. The feedback manager algorithm then repeats the test in another channel.

If a tonal signal or a whistling feedback signal is detected, the feedback manager algorithm automatically reduces the gain of the channel until the tonal signal disappears and sets a limit to the maximum allowable gain. The maximum gain reduction is achieved by adding another segment in the hearing aid input-output function around the lowest compression threshold (Figure 3A) (Brassine and Fabry, 2004, personal communication). The advantage of limiting the gain reduction at low level is that the gain and compression ratio of the mid-level or high-level inputs is not affected. The feedback manager algorithm then repeats the test in another channel (Fabry, 2004, personal communication).

Notice that the limit imposed by the optional feedback manager test sets the maximum gain allowable given the physical characteristics of the hearing aid. The feedback manager test should be rerun if the clinician made any modification to the shell or earmold, such as venting, lacquering of the hearing aid shell, switching to a thicker tubing, or replacing the earmold. Rerunning the test can remove the previous limits to maximum gain and set new limits, if any, for the new acoustic characteristics of the hearing aid or earmold. Should the removal of the gain limits be desirable without rerunning the feedback manager test, the clinician can deactivate the gain limit in the hearing aid fitting software.
Table 1. The Characteristics of Feedback Reduction Algorithms Implemented in Selected Commercially Available Hearing Aids*

<table>
<thead>
<tr>
<th>Method used to reduce feedback</th>
<th>Oticon Syncro</th>
<th>Phonak Perseo</th>
<th>ReSound Canta</th>
<th>Siemens Triano</th>
<th>Widex Diva</th>
</tr>
</thead>
<tbody>
<tr>
<td>Feedback detection and reduction</td>
<td>Phase cancellation</td>
<td>Adaptive gain reduction</td>
<td>Phase cancellation</td>
<td>Adaptive notch filters</td>
<td>Phase cancellation and gain reduction</td>
</tr>
</tbody>
</table>

1. A feedback path change detector monitors the gradual changes in the transfer function of the feedback path. The algorithm slowly adjusts the characteristics of the adaptive digital filter to approximate the transfer function of the feedback path and generates a cancellation signal. This internally generated signal is then subtracted from the microphone output to cancel the feedback path. This slow-acting mode of the adaptive filter cancels steady component of feedback and inaudible feedback signals that vary slowly.

2. A howl detector monitors the presence of tonal signals in the incoming signal. If a pure tone signal(s) is detected, the algorithm quickly adjusts the characteristics of the adaptive filter to approximate the feedback path and generates a cancellation signal. This internally generated signal is subtracted from the microphone output to cancel the feedback signal. The fast-acting mode of the adaptive filter cancels feedback due to abrupt changes in the acoustic environment and whistling feedback signals.

If the feedback cancellation algorithm fails to suppress feedback, the Feedback Manager test can be run in the fitting software. This test sets limits in the maximum gain allowable in the high-frequency channels while leaving the low-frequency channels unaffected.

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The feedback reduction algorithms of Perseo and Valeo also have an adaptive component that deals with abrupt changes in the acoustic environment. If feedback is detected in one or several frequency channels during daily use, the gains of the frequency channels are reduced by the same mechanism used in the feedback manager test (i.e., adding another segment in the input-output function around the lowest compression threshold). According to the manufacturer, this type of adaptive feedback management system has the advantage of lower power consumption than other adaptive feedback reduction algorithms (Fabry, 2004, personal communication). The disadvantage is that the gain of desirable signals may be reduced.

### 2.1.3.2. Adaptive Feedback Reduction by Notch Filtering

Notch filtering is another strategy to achieve feedback reduction. The general approach is to monitor the presence of tonal signals or whistling feedback signals. If a potential whistling feedback signal(s) is detected, the algorithm generates sharp notch filters at the feedback frequencies to suppress the whistling feedback signals. Algorithms from different manufacturers may differ in sharpness, depth, and number of notch filters they are capable of generating, the frequency range the filters operate in, and the speed of generating these filters, among others. In some algorithms, notch filters with different sharpness and depth are available. Sharp notch filters are generally used to limit the gain reduction at the feedback frequencies. The sharper the filter, the narrower the gain reduction region is and the less the effect on the gain of the hearing aid at the applied frequency region.

One example of this type of feedback reduction algorithm is found in Siemens Triano hearing aids (Table 1). The algorithm can generate four notch filters, two with a 6-dB notch and the other two with a 12-dB notch. It is manually activated by running a feedback manager test. During the feedback manager test, combination tones are presented across frequencies. If whistling feedback signals are detected, the algorithm inserts a notch filter with a 6-dB notch into the signal path with the center frequency matching the peak frequency of the feedback. If feedback is not totally suppressed in the first attempt, the algorithm switches to a notch filter with a 12-dB notch to reduce the feedback signal. This mechanism acts on the static component of feedback.

The adaptive component is activated by abrupt changes in the acoustic environment after the feedback manager test is finished. Its feed-
back reduction action is accomplished by the remaining notch filters that are not activated by the feedback management test. Two minutes after the notch filter is inserted, the algorithm removes it to check if the whistling feedback signal is still present. If present, the filters are reactivated until the instrument is switched off or another listening program is selected (Powers, 2004, personal communication; Siemens Audiology Group, 2004).

It is important for feedback reduction algorithms that use the notch filtering strategy to be able to generate multiple notch filters so that multiple simultaneously occurring whistling feedback signals can be reduced without generating the frequency-hopping artifact. If an algorithm can only generate a few notch filters or only a few notch filters are left to act on abrupt changes in the environment, a frequency-hopping artifact is generated when the number of simultaneously occurring whistling feedback signals exceeds the number of notch filters (Agnew, 1996).

To illustrate this problem, assume that there are three whistling feedback signals (S1, S2, and S3) and only two notch filters (F1 and F2). While F1 is suppressing S1 and F2 is suppressing S2, the feedback detector tells the algorithms that S3 is present. In response, the algorithm adjusts the frequency and amplitude characteristics of F1 to suppress S3. As soon as the adjustment is made, the feedback detector reports that S1 is present and the algorithm adjusts the characteristics of F2 to suppress S1. As there is always an uncovered whistling feedback signal, the algorithm adjusts one of the two filters to cover it, hence the filters hop among the feedback frequencies and a frequency-hopping artifact is generated.

2.1.3.3. Adaptive Feedback Reduction by Phase Cancellation

Phase cancellation is the third adaptive approach currently used to reduce feedback. Feedback reduction algorithms that use the phase cancellation strategy monitor the transfer function of the feedback path, generate a signal with a transfer function similar to that of the feedback path, and subtract the generated signal from the output of the hearing aid microphone. Another way to explain this process is that the algorithm modifies the characteristics of the digital filter dedicated for feedback cancellation so that its transfer function has similar frequency-amplitude-phase relationship as that of the feedback path. Then the output of this filter is subtracted from the microphone output to cancel feedback. Because of this canceling mechanism, these algorithms are often referred to as feedback cancellation algorithms.

Feedback cancellation algorithms have been implemented in hearing aids in many ways. An earlier version of an adaptive feedback cancellation algorithm estimates the transfer function of the feedback path by injecting a low-level noise
into the receiver and cross-correlating the receiver input with the microphone output. Then the algorithm modifies the digital filter characteristics to create a cancellation signal (Egolf, 1982; Engebretson and French-St. George, 1993; Kates, 1991; Danalogic, 2000; Kuk et al., 2002; Olsen et al., 2001; Maxwell and Zurek, 1995). However, many hearing aid users found the continuous low-level noise annoying. Thus, this type of feedback cancellation algorithm is only used in hearing aids fitted to people with severe hearing loss.

Newer versions of the feedback cancellation algorithm eliminate the noise injection and reduce feedback in multiple signal processing stages. Specific implementations of feedback cancellation algorithms are found in the GNResound Canta and Air, Oticon Syncro and Adapto, among others.

The feedback cancellation algorithm in GNResound Canta and Air has a fixed filter and an adaptive filter. The transfer function of the fixed filter is determined in a feedback path estimation test during the hearing aid fitting session. During the feedback test, a noise burst is generated internally and sent to the receiver (Figure 4A). Then the noise burst sent to the receiver is cross-correlated with the microphone output of the noise burst with multiple delay times to estimate the transfer functions of multiple feedback paths. These feedback paths include sounds leaking out from the receiver to the microphone via the vent, the sides of the earmold or shell, the tubing wall, and sounds transmitted to the microphone via the faceplate of the shell, inside the hearing aid (Hellgren et al., 1999a; Thompson, 2004 personal communication). This cross-correlation process allows the algorithm to estimate the summation of the transfer function of multiple feedback paths when the hearing aid is worn in the user’s ear. An example of the estimated transfer function of the feedback path is shown in Figure 2. This fixed filter accounts for the static component of feedback given the characteristics of the hearing aid and the user.

After estimating the transfer function of the feedback path, the algorithm modifies the characteristics of the fixed filter to approximate or model the transfer function of the feedback path and generates a cancellation signal. Examples of the transfer functions of the feedback path and the cancellation signal are shown in Figure 4B. Notice that the feedback cancellation algorithm cancels the feedback path even in the absence of any periodic, tonal, or feedback signal (i.e., it cancels the feedback path of the inaudible feedback signal) (Edwards, 2004, personal communication; Kates, 2003; Thompson, 2004, personal communication).

If there are enough poles (peaks), zeros (dips) or filter coefficients, this cancellation process is capable of canceling feedback at multiple frequencies. Poles, zeros, and filter coefficients are parameters to define the characteristics of the digital filter. The more poles, zeros or filter coefficients, the more peaks and dips can be generated in the transfer function of the digital filter and the better is the digital filter to approximate the feedback path (Thompson, 2004, personal communication). The limitations of the fixed filter are that its characteristics are fixed after the fitting session and it cannot act on the dynamic component of feedback.

To overcome these limitations of the fixed filter, an adaptive filter is implemented in the Resound Canta to act on feedback that is generated by the dynamic component of feedback (Figure 4C). The feedback cancellation algorithm adopts the transfer function of the fixed filter as the baseline and constantly performs cross-correlation between the receiver input and the microphone output to monitor the changes in the transfer function of the feedback path. Should a whistling feedback signal or a change in the transfer function of the feedback path be detected, the algorithm changes the characteristics of the adaptive digital filter to approximate the transfer function of the new feedback path. Then a signal with the combination effect of the fixed and adaptive filters is internally generated and subtracted from the microphone output to cancel feedback (Groth, 1999; Kates, 1999; Olsen, 2001; Thompson, 2004 personal communication).

A different implementation of the feedback cancellation algorithms is found in hearing aids from Oticon (i.e., Syncro, Adapto, Gaia and Atlas). Instead of having two adaptive digital filters, these algorithms function in either the slow or the fast mode (Flynn, 2003; Oticon, 2004) (Figure 5). In addition, they are automatically turned on as hearing aids are turned on (i.e., they do not require a feedback path estimation test to be activated).

The algorithms use two detectors, a howl detector and a feedback change detector, to monitor changes in the feedback path at frequencies above 1500 Hz. As mentioned before, acoustic feedback is most likely to occur between 2-5 kHz. The high-
Figure 4. The feedback cancellation system implemented in ReSound Canta. (A) The diagram of the fixed digital filter implemented in the feedback cancellation algorithm of ReSound Canta. The transfer function of the feedback path is approximated in several steps: (1) An internally generated probe signal is sent to the receiver; (2) the probe signal leaks out via various feedback paths and is picked up by the microphone; (3) the input of the receiver is cross-correlated with the output of the microphone and the transfer functions of the feedback paths are estimated; (4) the algorithm adjusts the characteristics of the static digital filter to approximate the resultant transfer function of the feedback paths. Reprinted and modified with permission from Groth (1999), Hearing Journal, 52(5). (B) The transfer function of the feedback path (solid line) and the transfer function of a digital filter (dash line) approximating that of the feedback path. Notice that the two transfer functions are not identical. Reprinted with permission from Kates (2003), Journal of Acoustical Society of America, 106. (C) A diagram of the adaptive and fixed digital filter implemented in the feedback cancellation algorithm of ReSound Canta. The algorithm uses the characteristics of the static filter as a baseline and constantly monitors the changes in the transfer function of the feedback path. If a change in the feedback path or a whistling feedback is detected, the algorithm modifies the characteristics of the adaptive filter and generates a cancellation signal. This signal is then subtracted from the microphone output to cancel feedback. Reprinted and modified with permission from Groth (1999), Hearing Journal, 52(5).
pass filter was added in the signal path to avoid canceling tonal sounds with frequencies below 1500 Hz (Flynn, 2004, personal communication).

The feedback change detector is used to detect gradual changes in the transfer function of the external feedback path. When a change is detected, the algorithm gradually varies the characteristics of the adaptive filter to generate a cancellation signal to cancel the external feedback path. The algorithm normally operates in the slow mode to conserve battery and reduce signal processing demand (Flynn, 2003). If the howl detector detects potential whistling feedback signals at the microphone output, the algorithm quickly modifies the characteristics of the adaptive filter and generates a cancellation signal to cancel the tonal signal.

If the feedback cancellation algorithm is unable to totally suppress whistling feedback signals, a feedback manager test can be run in the fitting software during the fitting session. This test sets limits of the gains in high-frequency channels to manage feedback but leaves the gains in low-frequency channels unaffected (Flynn and Flynn, 2004).

2.1.3.4. Combination of Feedback Cancellation and Adaptive Gain Reduction

The previous section discussed the three primary strategies currently used to reduce feedback. Some manufacturers have combined different strategies in their adaptive feedback reduction algorithms. For example, Widex Senso Diva uses both phase cancellation and adaptive gain reduction strategies to control feedback. The adaptive feedback reduction algorithm in Diva is composed of three components: a fixed filter, a slow-acting adaptive filter, and a fast-acting gain reduction component. Both fixed and slow-acting adaptive filters use the phase cancellation strategy, and the fast-acting component uses the adaptive gain reduction strategy.
The characteristics of the fixed filter in Diva are defined in a feedback test in a process similar to that in GNReSound Canta. Its characteristics are fixed after the feedback test is run and it accounts for user and hearing aid characteristics. Diva uses the combination of a slow-acting adaptive filter and a fast-acting gain reduction component to reduce the dynamic component of feedback. Specifically, the algorithm uses the parameters of the fixed filter generated in the fitting session as the baseline and constantly monitors the feedback path. When gradual changes in the transfer function of the feedback path are detected, the algorithm adjusts the transfer function of the slow-acting adaptive filter and generates a cancellation signal. When a tonal signal is detected in the microphone output, the algorithm quickly reduces the gain for low-level inputs at the frequency channel in which the tonal signal occurs. Then the algorithm gradually changes the transfer function of the slow-acting adaptive filter to approximate that of the new feedback path, and the fast-acting gain reduction component gradually releases the amount of gain reduction (Kuk et al., 2002a).

The purposes of the fast-acting gain reduction component are to reduce whistling feedback signals quickly through gain reduction and to allow time for the slow-acting adaptive filter to approximate the transfer function of the new feedback path. In contrast to the gain reduction strategy used in Phonak Perseo, Diva achieves gain reduction by increasing the compression threshold of the frequency channel (Figure 3B). The advantage is similar to that of the Perseo: the gain at low level is reduced but the compression ratio and the gain at middle and high levels are not affected.

Another function of the fast-acting gain reduction component is to compensate for the limitations of the fixed and slow-acting adaptive filters. If the combination of the two adaptive filters is unable to totally cancel the feedback signal or the algorithm is unable to estimate a stable transfer function of the feedback path because of continuous changes in the feedback path, the fast-acting component maintains the amount of gain reduction that is necessary to control feedback (Kuk et al., 2002a).

The slow-acting adaptive filter in Diva has an attack time of about 5 to 10 seconds and the attack time of the fast gain-reduction component is less than 1 second. A short time constant is used for the fast gain-reduction component to reduce feedback as soon as feedback is detected. The slow-acting adaptive filter has a relatively long time constant because it uses a relatively long signal detection and analysis window to estimate the transfer function of the feedback path to reduce artifacts (Kuk et al., 2002a).

2.1.4. Advantages of Adaptive Feedback Reduction Algorithms

Adaptive feedback reduction algorithms have many advantages. Algorithms that use adaptive gain reduction and notch filtering strategies only reduce gain or insert notch filters when a whistling feedback signal is detected. Algorithms that use the phase cancellation strategy constantly monitor changes in the transfer function of the feedback path and generate cancellation signals with transfer function approximating the transfer function of the feedback path to cancel feedback. Depending on the parameters kept constant, manufacturers and clinicians have applied these adaptive feedback reduction algorithms to achieve several fitting goals.

If the venting option is kept the same, activating feedback reduction algorithms can increase headroom, the maximum gain allowable, and the fitting range of the hearing aid. Feedback reduction algorithms can be especially useful if the gain is set just below the level at which feedback occurs. Activating these algorithms can improve speech intelligibility and sound quality by eliminating the sub-oscillatory feedback (Cox, 1982; Latzel et al., 2001).

Adaptive feedback reduction algorithms also change the behaviors of clinicians and hearing aid manufacturers. Because of the increase in the allowable gain, clinicians can now fit clients with smaller custom hearing aids (e.g., a client previously recommended for half-shell in-the-ear hearing aids may be able to be fitted with in-the-canal hearing aids). Manufacturers are also known to increase the recommended fitting range of their hearing aids because of the availability of adaptive feedback reduction algorithms.

On the other hand, if the gain is kept constant, activating adaptive feedback reduction algorithms allows the user to have a larger vent, use the telephone, or wear a hat without permanently reducing the gain of the hearing aid. Adaptive feedback reduction algorithms are the main contributors of open ear canal fittings. One manufacturer reported that its feedback cancella-
tion algorithm allowed an increase of vent size from 1.7 mm to 3.2 mm (Flynn, 2003). A larger vent reduces the occlusion effect and consequently improves the sound quality when the hearing aid user is speaking, chewing, swallowing, or even walking. The ultimate goal of adaptive feedback reduction algorithms is to completely open up the ear canal for listening comfort.

2.1.4.1. Verification
The verification data of adaptive feedback reduction algorithms span a wide range. Laboratory tests reported about a 10 to 35 dB of increase in usable gain (Dai and Hou, 2004; Dyrlund and Bisgaard, 1991; Egolf, 1982; Hayes and VonLanthan, 2003; Siqueira and Alwan, 2000). However, room acoustics and currently available digital signal processing power in digital hearing aids limit the effective range of feedback cancellation algorithms to about a 5 to 15 dB increase of usable gain in field studies (Flynn and Flynn, 2004; Kuk et al., 2002; Mueller, 2002; Smirga and Groth, 1999; Stone and Moore, 2002; Siemens Audiology Group, 2004). It should be noted that only when adaptive feedback reduction algorithms are used to increase headroom, can the occurrence of feedback be significantly reduced.

If these feedback reduction algorithms are applied to increase the vent size or the fitting range, a hearing aid may still generate feedback or experience significant gain reduction when the acoustic environment changes. Under these conditions, the feedback reduction algorithms are already engaged in feedback reduction so that when abrupt changes occur in the acoustic environment, they may not be able to meet further demand to reduce feedback. In algorithms in which gain reduction is applied to control feedback, further gain reduction is applied in response to the abrupt changes in the acoustic environment, which may result in loss of audibility to desirable sounds.

2.1.5. Effect of Algorithms Using Different Strategies on Hearing Aid Output
Harris and colleagues (Harris, 2004, person communication) have examined the effect of several feedback reduction algorithms on the level of speech sounds at the hearing aid output. They included four hearing aids from three manufacturers in the study:

- one hearing aid using conventional multiband gain reduction strategy (7 bands),
- one hearing aid using adaptive notch filtering strategy, and
- two hearing aids using adaptive phase cancellation strategy.

Their goal was to determine if these algorithms significantly reduce the target speech signal (an undesirable effect) while they are in action. All hearing aids were tested under two conditions:

1. The overall gain of the hearing aid was set 1 dB below the level at which a whistling feedback signal is detected when the feedback reduction algorithm was not activated, and
2. The overall gain of the hearing aid was increased 5 dB above the setting in condition 1 and the feedback reduction algorithm was activated. The output of the hearing aid recorded by a probe microphone was reduced by 5 dB to compensate for the gain increase.

Consonant-vowel-consonant nonsense syllables were presented at 55 dB SPL and recorded at the output of the hearing aids. When the outputs were compared in the two conditions, Harris and colleagues found that speech processed by the hearing aids using adaptive feedback cancellation algorithm maintained the overall level of the speech sounds in condition 2. However, the level of speech processed by the feedback reduction algorithms using the gain reduction and the notch filtering strategies was reduced by an average of 8 and 4 dB, respectively.

These results indicated that although no permanent gain reduction, any adaptive feedback reduction algorithms that use the gain reduction and notch filtering strategies may reduce the audibility of desirable signals when they are engaged. In addition, feedback cancellation is the only feedback reduction strategy that does not affect the gain or output of the hearing aid and therefore maintains the audibility of speech. These findings are consistent with previous laboratory tests of feedback cancellation algorithms (Dyrlund and Bisgaard, 1991; Engebretson and French-St. George, 1993; Kates, 1991; Kates, 2003).

Harris and colleagues also measured the overall gain level achieved before and after the activation of the feedback reduction algorithms. Their results showed that the activation of any algorithms...
type of feedback reduction algorithm allowed additional gain to be programmed in all the test hearing aids.

A point worth noting is the difference between the output of the algorithms using the gain reduction and the notch filtering strategies. Some feedback reduction algorithms that use a gain reduction strategy only reduce the gain for low-level inputs, and the gain and compression ratios are not affected at middle and high levels. However, if gain reduction is applied across the input levels or if notch filtering is used, the output of the hearing aid can be affected at all input levels (Figure 6).

2.1.6. Challenges and Limitations
There are many challenges and considerations in the design of feedback reduction algorithms. Depending on the particular implementation, these challenges may impose limitations on the effectiveness of feedback reduction algorithms. If these challenges are not adequately addressed, undesirable artifacts may be generated. The following section discusses several challenges in the implementation of feedback reduction algorithms.

2.1.6.1. Factors Limiting the Performance of Feedback Reduction Algorithms
Two factors that limit the performance of adaptive feedback reduction algorithms in the real world are the available space for the digital signal processing (DSP) chip and the battery supply in hearing aids. As cosmetics is one of the biggest concerns of hearing aid users, the size of the DSP chip is limited to an area of less than 1 cm². For a hearing aid to yield at least 50 hours of battery life on a 1.3 to 1.5-V supply as demanded by the market, the current drain from the button battery is normally limited to 0.5 to 1.5 mA. These size and power limitations result in the use of DSP chips with low processing speed, which in turn limits the complexity of the signal-processing algorithms that can be implemented in wearable hearing aids (Kates, 2003).

Figure 6. The frequency responses of a hearing aid with notch filters on and off at different input levels. The differences between traces 1 and 4, traces 2 and 5, and traces 3 and 6 in the high-frequency region are the amount of gain reduction introduced by the notch filters.
In theory, if an adaptive feedback reduction algorithm could have unlimited DSP power and could draw power from wall socket, it could suppress or cancel feedback with any amplitude. In reality, being implemented in DSP chips with relatively low processing speed compared with central processing units in desk top computers and being restricted by the allowable processing delay, adaptive feedback reduction algorithms are restricted on their speed and ability to reduce gain, generate notch filters, or model the transfer function of the feedback path.

Kates' (2003) review specifically discussed factors affecting the effectiveness of feedback cancellation algorithms caused by the above limitations. The ability of the feedback cancellation algorithms to accurately approximate or model the transfer function of the feedback path is determined by the number of coefficients of the digital filter and whether a combination of poles (peaks) and zeros (dips) is used in feedback cancellation. With a higher number of filter coefficients and more poles and zeros, the feedback cancellation filter can model a higher number of peaks and dips and steeper slopes, which results in better cancellation of the feedback signals (Kates and Thompson, 2004, personal communication).

Another factor limiting the performance of feedback cancellation algorithms is reverberation. Room reverberation affects the ability of feedback cancellation algorithms, because it consists of multiple acoustic reflections with different amplitudes and time delays. It causes peaks and valleys in the transfer function of the feedback path that can be difficult to approximate with limited signal processing power and a limited number of filter coefficients, poles, or zeros available to the algorithms (Kates, 2003). Thus, the magnitude and number of feedback peaks that the algorithms can successfully cancel is limited.

Space, current drain, signal processing speed, reverberation, and the complexity of the algorithm limit the maximum feedback cancellation ability to 5 to 15 dB in the real world (Kates, 2003). As technologies in chip design advance further, it is expected that future generations of adaptive feedback reduction algorithms will be more sophisticated and capable of canceling feedback with a higher magnitude and a higher number of feedback peaks.

2.1.6.2. The Frequency-Hopping Artifact and the Chirping Artifact

The number of notch filters or the ability of the feedback cancellation algorithm to approximate the transfer function of the feedback path determines the ability of the algorithm to reduce multiple simultaneous feedback signals and thus, determines the number of occurrence of the frequency-hopping artifact. It is important to mention that this frequency-hopping artifact is not limited to feedback reduction algorithms that use the notch filtering strategy. Feedback cancellation algorithms with limited ability to model the transfer function of the feedback path can also generate this artifact (Kates and Nilsson, 2004, personal communication).

Recall that feedback cancellation algorithms have filters to approximate the transfer function of the feedback path and generate a signal to be subtracted from the microphone output to cancel feedback. The frequency-hopping artifact in a feedback cancellation algorithm occurs when the feedback cancellation filter keeps adapting to the feedback signal that is not adequately canceled because the generated cancellation signal cannot effectively model the transfer function of the feedback path.

In theory, the three-component feedback reduction algorithm implemented in the Widex Diva can reduce the frequency-hopping artifact because the fast gain-reduction component maintains the amount of gain reduction necessary to suppress feedback if the slow adaptive component is unable to model the transfer function of the feedback path and to suppress feedback. This practice may reduce the gain for desirable signals, however.

Another artifact associated with feedback cancellation algorithms is the “chirping” artifact. If an algorithm makes a mistake in the estimation of the transfer function of the feedback path, it generates and injects a “cancellation signal” at the microphone output yet there is no feedback signal to cancel. This injected cancellation signal is sometimes perceived as chirping by the hearing aid user (Kates, 2003).

2.1.6.3. Differentiating Feedback Signals and Musical Notes

Another great challenge in the implementation of the adaptive feedback reduction algorithm is the ability of the algorithm to recognize desirable tonal signals such as musical notes, microwave beeps, or other desirable tones, and leave them...
unaltered. These signals are composed of pure tones that have similar acoustic characteristics as feedback signals. Depending on the particular implementation of the feedback reduction algorithm, a sustained or a brief musical note may be mistaken as a whistling feedback signal and canceled.

Different feedback reduction algorithms may use different strategies to avoid this mistake. Some feedback reduction algorithms impose a limit on the frequency range that the feedback detectors monitor or the notch filters can be applied. For examples, the feedback detector of Oticon Syncro and Siemens Triano only monitor feedback signal(s) at frequencies above 1500 Hz or 2000 Hz, respectively (Flynn at Oticon, 2004, personal communication; Hamacher at Siemens, 2004, personal communication). This limit can avoid identifying musical notes with frequencies below these cut-off frequencies to be treated as feedback.

Another strategy feedback reduction algorithms may use to differentiate feedback from musical notes is to use a long signal-detection and analysis window (e.g., 10 seconds), pattern recognition to analyze the pattern of the incoming signal, and slow adaptation time constants for the adaptive filter (Kuk, 2004, personal communication, Kuk et al., 2002). Within this analysis window, the detection and analysis unit analyze the frequency, spectral, or temporal pattern of the incoming signal and try to determine if the signal is a musical or feedback signal. In addition, slow adaptation time constants are used for the adaptive filter so that short tonal signals are not canceled.

In theory, these features can reduce the chance of the feedback reduction algorithm acting on musical notes in the middle of a musical piece. It is unknown if this algorithm can differentiate the first few musical notes from whistling feedback signals (e.g., when a violinist tunes the violin before playing a musical piece).

Another practice to avoid canceling desirable tonal signals is to restrain the amount of filter adaptation in the feedback cancellation algorithm (Kate, 2003). In this approach, the algorithm constantly compares the updated feedback path transfer function with the baseline feedback path transfer function. If the two transfer functions deviate for more than a predetermined amount, the adaptation of the transfer function is constrained. The result of such a constraint is that the algorithm cannot accurately model the transfer function of the feedback path, and therefore, the tonal signal is not completely canceled (Kates, 2004, personal communication). For detailed description, please see Kates (2003).

The general practice for working with most feedback reduction algorithms (and noise reduction algorithms) is to turn the algorithms off for music enjoyment. When fitting a hearing aid with feedback reduction algorithm, clinicians need to discuss the characteristics of the feedback reduction algorithm with the hearing aid user and reserve a program for music enjoyment. As feedback reduction algorithms are, sometimes, used to expand the fitting range or open a larger vent, it is important to make sure that the hearing aid does not have feedback when the feedback reduction algorithm is turned off.

Recently, Siemens has launched a new model of hearing aid, Acuris, and claims that the feedback cancellation algorithm in Acuris can distinguish environmental tonal signals and feedback signals 75–80% of the time (Powers, 2004, personal communication). The exact mechanism is proprietary at this time.

2.1.6.4. Open Ear Canal Fitting
Feedback reduction algorithms have become an integral part of open ear canal fitting. They allow a larger vent to be used to reduce the occlusion effect without risking feedback. Several challenges accompany open ear canal fittings because of the vent effect, and among them is the need for hearing aids to have short processing delays. When a hearing aid has a large vent, unprocessed sounds in the environment (especially the low-frequency contents) can go into the ear canal through the vent. If there is a perceivable mismatch between sounds processed by the hearing aid and unprocessed sounds entering the ear canal through the vent, the user may find the experience objectionable.

Another challenge of open ear canal fitting arises from the interactions of sounds getting into and leaking out from the ear canal. Sounds in the environment can get into the ear canal after being processed by the hearing aid, and unprocessed sounds can get into the ear canal via the vent (Figure 7A). Sounds in the ear canal can also leak out from the vent. As shown in Figure 7B, sounds leaking out of the vent have a high-pass filter characteristic, and sounds getting into the vent have a low-pass filter characteristic (Scheller, 2004). If the gain of a hearing aid at the low-frequency region is not very high, unprocessed sounds
can potentially mask the processed sounds. Consequently, the directivity of directional microphones, and the effectiveness of noise reduction algorithms are reduced.

Figure 7B shows the vent effect of a 2.3-mm vent. If the vent size increases, the cut-off frequencies of sounds leaking out of and getting into the vent increase (*i.e.*, more processed sounds leak out from the ear canal and more unprocessed sounds get into the ear canal). In addition, the net low-frequency gain provided by a hearing aid also decreases as the vent size increases. To lessen the negative effects of a large vent, some manufacturers automatically increase low-frequency
gains according to the vent size chosen in the fitting software. Clinicians also should be vigilant not to use excessively large vents for hearing aid users with significant low-frequency hearing loss.

The phase relationship between the processed and unprocessed sounds in the ear canal can also pose a challenge in open ear canal fitting, especially in the frequency regions where the unprocessed sounds and the processed sounds have comparable amplitudes. From basic principles of physics we know that if two sounds with identical frequencies and amplitudes (say 500 Hz at 65 dB SPL) are added, the resultant sound pressure level can vary from 0 dB SPL to the original level plus 6 dB, depending on the phase relationship between the two sounds.

Specifically, if the two sounds are 180° out of phase, the resultant sound pressure level equals 0 dB SPL. If the two sounds are in phase, the resultant sound pressure level is 6 dB higher than the original level \( i.e., 65 + 6 = 71 \text{ dB SPL} \). Intermediate levels can result as the phase relationship between the unprocessed and processed sounds varies between 0° to 180°. In hearing aid fitting, the frequency response measured in the ear canal may show a peak or a dip at the frequency region where the unprocessed sounds have similar amplitudes as the processed sounds.

The processing delay of digital hearing aids may also make the matter worse. With processing delay, the processed sounds lag the unprocessed sounds by several milliseconds. The effect of this delay is analogous to two signals with a phase difference. According to Scheller (2004), sounds processed by digital hearing aids have more rapid variations in phase across frequencies than unprocessed sounds getting into the ear canal via the vent. This can result in multiple peaks and dips in frequency regions where the processed and unprocessed sounds have comparable amplitudes (Scheller, 2004). Fortunately, this erratic effect is negligible in frequency regions where sounds coming from these two sources have larger level differences because the characteristics of the dominant sound source determine that of the resultant sounds in the ear canal.

### 2.1.7. Working with Adaptive Feedback Reduction Algorithms

When working with hearing aids with feedback reduction algorithms, clinicians need to be aware of several issues:

1. Feedback reduction algorithms need to be switched off when any electroacoustic test using pure tone signals is conducted. Otherwise, the feedback reduction algorithm treats the test signal as feedback, tries to suppress it, and generates inaccurate results.

2. Clinicians need to inform hearing aid users that the feedback signal emitted by hearing aids with adaptive feedback reduction algorithms may be different from the conventional high-frequency whistling feedback signals that they are used to hearing. The reason is that, instead of a constant pure tone signal, the frequency-hopping artifact generated by algorithms with adaptive notch filters or by feedback cancellation algorithms may vary in frequency and intensity.

3. If adaptive feedback reduction algorithms are used to achieve open ear canal fitting or to increase the fitting range of hearing aids, clinicians need to be careful to avoid the vent size and/or the gain level close to where feedback begins to occur. Otherwise, the hearing aid becomes very prone to feedback when abrupt changes in the environment occur.

4. Sub-oscillatory feedback causes multiple peaks in the frequency response and degrades the sound quality of speech (Cox, 1982; Langford-Smith, 1960). Clinicians also need to be aware that if the hearing aid user suspects the hearing aid is malfunctioning, the degradation could be due to sub-oscillatory feedback (Danalogic, 2000). A good clinical practice is to leave 4 to 8 dB of headroom to avoid these negative effects (Skinner, 1988).

5. For hearing aids with feedback manager or feedback path estimation test, it is a good practice to run these tests periodically after the initial fitting session. Performing the tests can take the changes in the ear canal over time into account and renew the baseline settings of the feedback reduction algorithm or the maximum allowable gain of the hearing aid. It is also very important to perform these tests every time physical characteristics of the hearing aid shell or earmold are modified.

An accurate feedback manager test or feedback path estimation enhances the effectiveness of feedback reduction algorithms. In addition, it reduces the activity of the adaptive or fast-acting component of the algorithms and thus increases the stability of the algorithms and reduces battery consumption. Further,
some feedback reduction algorithms impose a limit on the maximum gain allowable after running the initial test. If additional gain is needed, clinicians can modify the physical characteristics of the hearing aids, for example, changing the vent size or adding an extra coat to the custom shell. Rerunning the test can release the limit set on the maximum gain and allows higher gain settings. Otherwise, this baseline needs to be reset manually in the fitting software.

6. The availability of the feedback reduction algorithm should not undermine or preclude the application of other solutions to resolve feedback problems in a clinical practice, including for example, a properly fitted earmold or custom shell, an appropriate vent size, a properly oriented receiver in the ear canal, an intact tubing with properly sealed joints, a cerumen-free receiver tubing, no excessive cerumen in the ear canal, a proper hearing aid style (Kuk et al., 2002; Olsen et al., 2001).

In fact, some fitting software gives out a warning message if the amount of gain limit needed to control feedback exceeds the limit of the algorithm or exceeds a certain number of dB. For example, during the feedback manager test, the Siemens Triano fitting software warns the clinician if the amount of gain reduction needed to control feedback is more than 12 dB. It also advises the clinician to consider other feedback management options before proceeding with the hearing aid fitting.

2.2. Mechanical and Electromagnetic Feedback

Mechanical feedback occurs when sounds caused by the vibration of the receiver are transmitted to the microphone via the shell and the faceplate of custom hearing aids or when sounds caused by the vibration of tubing are transmitted to the microphone. The former is caused by the force generated by in-phase movements of the paddle and the reed inside a conventional receiver.

In a conventional receiver, a motor is formed by a U-shaped reed, a coil, a spacer, two magnets and a yoke of magnetic material that confines the magnetic field and returns it to the magnets (Figure 8A). When current flows in the coil, a magnetic field is generated in the reed. The displacement of the reed from the resting position is proportional to the current flowing in the coil. A straight, rigid driving rod connects the reed to the paddle and is used to transmit the movements of the reed to the diaphragm, which vibrates air and produces sound. The movement of the paddle and the reed create a net reaction force that rocks the case of the receiver (Figure 8B). If this mechanical force is transmitted to the shell and then to the faceplate of the hearing aid, it vibrates the air close to the faceplate and the microphone can pick up the air pressure change. If the air pressure change is strong enough, it may cause feedback (Agnew, 1996; Thompson, 2002).

Magnetic feedback is caused by the stray magnetic field generated by the current flowing in the coil of the receiver. Sometimes, this stray magnetic field is picked up by the telecoil and causes magnetic feedback.

2.2.1. Mechanical and Magnetic Feedback Reduction Strategy: Redesigned Receiver

Recently, receivers have been redesigned with two new features to reduce mechanical and magnetic feedback. In the redesigned receivers, the straight, rigid driving rod is replaced by a diamond-shaped metal part known as a pantograph. When the reed moves upward, the arms of the pantograph move outward to the sides, pulling down the paddle (i.e., the reed and the paddle move toward each other) (Figure 8C). When the reed moves downward, the arms of the pantograph move in from the sides, pushing the paddle upward (i.e., the reed and the paddle move away from each other). In other words, the reed and the paddle move in opposite directions and the action and reaction forces cancel each other (Figure 8D).

As little net force (in theory) is generated by the movements of the reed, the paddle, and the pantograph, feedback due to mechanical force is greatly reduced. The transmission of mechanical force to the faceplate is further reduced by adding a layer of rubber padding on each end of the receiver. To reduce magnetic feedback, a metal can is placed outside of the rubber-padded receiver to create an electromagnetic shield and to reduce the intensity of the stray magnetic field leaking to the telecoil.

2.2.2. Verification

The redesigned receiver, in theory, should reduce the amount of mechanical and magnetic feedback. Few published verification data are available at this time to evaluate its effectiveness.
Figure 8. (A) The components of a hearing aid receiver. (B) A force is generated at the reed when an electrical signal reaches the receiver. This force is transmitted to the paddle via a rigid driving rod which generates a force in the same direction as the force generated in the reed. A reaction force to the opposite direction is generated at the receiver case, creating a rocking movement. Feedback is generated if this rocking energy is transmitted to the faceplate, which vibrates the surrounding air and is subsequently picked up by the microphone. (C and D, following page) The newly designed diamond-shaped pantograph reduces mechanical feedback by generating reaction forces in the opposite direction to the driving forces. Figures provided by courtesy of S. Thompson at Knowles Electronics.
3. Challenge No. 3.
The Occlusion Effect

The occlusion effect is one of the most frequent complaints of hearing aid users. It was first described by Zwislocki (1953) that during a bone conduction test, more sounds were produced in the ear canal if an earplug was placed in the cartilaginous portion of the ear canal than if the earplug was placed in the bony portion of the ear canal. Later, Khanna and colleagues (1976) used a probe microphone to measure the sound pressure between the eardrum and the earplug. They confirmed that the occlusion effect was generated...
when the energy of a bone-conducted sound vibrates the cartilaginous portion of the ear canal and sounds generated by the vibration cannot escape from the ear canal.

In hearing aid usage, the occlusion effect is generated when a hearing aid or an earmold is worn in the cartilaginous portion and occludes the ear canal. The bone-conducted sounds generated during vocalization, chewing, swallowing, or walking are trapped inside the ear canal. As the occlusion effect is most prominent at frequencies below 500 Hz, it offsets the natural level differences between the vowels and consonants and creates a booming sensation when the hearing aid user vocalizes. When a hearing aid with no vent is worn, the low-frequency energy of vowels can be as much as 20 to 30 dB louder than the same vowels in an unoccluded in ear canal (Gudmundsen, 1994; Killion et al., 1988; Kampe and Wynne, 1996; Mueller et al., 1996; Revit, 1992). This excessive low-frequency energy may mask the consonants before (i.e., backward masking) or after (i.e., forward masking) the vowels in one's own speech.

Because of this low-frequency dominance, the occlusion effect is often most bothersome to hearing aid users with normal or nearly normal low-frequency hearing. Hearing aid users with more low-frequency hearing loss may find the occlusion effect less annoying because it is less likely for them to hear low-frequency sounds. In addition, their hearing aids typically have more low-frequency gain. The more pleasant amplified sounds may mask the unpleasant occluded sounds (Mueller, 2003).

The amount of the occlusion effect is usually measured by using a probe microphone placed in the ear canal. The measurement is accomplished in two steps:

1. The sound pressure level of the hearing aid user's own voice during a vocalization of an /i/ or /u/ is measured in open ear canal.
2. The sound pressure level of the same vowel vocalized at the same intensity as in (1) is measured when the hearing aid is turned off and placed in the ear canal.

The difference between these two measurements is the amount of occlusion effect. These measurements can be obtained using real ear measurement equipment, a probe microphone with its output fed to a spectrum analyzer (software or hardware) or an Occlusion Effect Meter (ER-33, Etymotic Research). The first two tools provide frequency-specific information on the occlusion effect. The Occlusion Effect Meter automatically calculates and reports the r.m.s. energy differences between 180 and 460 Hz in one reading.

It is important to note that not all booming sensation is created by the occlusion effect. Sometimes, the sensation is created by over-amplification in the low-frequency region or vent resonance effect (Mueller, 2003, Kuk and Ludvigsen, 2002c). If the former, the sensation can be reduced by reducing the low-frequency gain of a hearing aid. If the latter, the altering the vent size or vent length may change the characteristics of the resonance and thus reduce the booming sensation.

Conventional ways to manage the occlusion effect during the hearing aid fitting process include:

1. inserting the hearing aid or earmold into the bony portion to reduce the amount of the sounds generated in the ear canal;
2. increasing the vent size to let the sounds generated in the ear canal escape;
3. reducing the vent length;
4. increasing the low-frequency gain so that the amplified sounds can mask the undesirable occluded sounds; and
5. counseling.

Several hearing aid manufacturers have recently built on the first three concepts and launched hearing aids with new sound-delivery devices, modified configurations, and new vent designs. These approaches have significantly reduced the occlusion effect. The following section discusses the mechanism, advantages, and limitations of some of these approaches.

3.1. Strategy No. 1: Replacing the Earmold
To reduce the occlusion effect, several manufacturers have replaced the conventional ear molds of the behind-the-ear hearing aids with newly designed sound-delivering devices. For example, GNReSound has launched Air, its mini behind-the-ear hearing aids, for people with normal low-frequency hearing and high-frequency hearing loss. Air uses a relatively hard and thin (compared to the standard #13 tubing) preformed and presized tubing to deliver the sounds from the hearing aids to the ear canal (Figure 9A). A plas-
tic dome with holes on the sides and a sports bend are used to increase the tubing retention in the ear canal. The holes on the sides of the dome allow the low-frequency bone-conducted energy to escape from the ear canal and thus reduce the occlusion effect (Christensen, 2004, personal communication).

Phonak has recently launched a “Fit-and-Go” sound delivery device for its Perseo Open. The concept of the “Fit-and-Go” system is similar to that of the open fitting used in Air. Standard #13 tubing is used to transmit sounds from the behind-the-ear hearing aid to the ear canal (Figure 9B). A retention hook is used to hold the tubing in the ear. A soft tip is attached onto the tip of the tubing to increase comfort when the tubing is placed in the ear canal.

The fitting range for both Air and Perseo Open is normal hearing at the low-frequency region and up to 70 dB HL of hearing loss at the high-frequency region. For people with more hearing loss, a more occluding sound-delivering device (i.e., an earmold) or an in-the-ear hearing aid with a vent is needed. Additional information on these two hearing aids is listed in Table 2.

Since the initial preparation of this manuscript, Widex has launched Elan with narrow tubing and an open sound-delivery device. It is expected that other manufacturers will also launch new sound-delivering devices with a similar configuration and function in the near future.

3.1.1. Verifications and Limitations
The open ear canal occlusion effect measured by the Occlusion Effect Meter (ER-33) is about −2.5 dB (Staab, 2003). GNReSound reported that the occlusion effect measured on subjects wearing Air was −2.5 to −5 dB (Christensen, 2004). This result indicated that the open-fitting sound-delivery device virtually eliminated the occlusion effect associated with hearing aid use. Perseo was launched in the summer of 2004. No published validation data on its performance are available to date.

These open-fitting sound-delivering devices have advantages as well as limitations. They are cosmetically appealing and can totally open up the ear canal. However, they provide limited low-frequency amplification. This limits the fitting range to normal or nearly normal hearing sensi-
### Table 2. The Characteristics of Commercially Available Hearing Aids With Novel Sound Delivering Devices to Reduce the Occlusion Effect

<table>
<thead>
<tr>
<th>Component</th>
<th>Phonak Perseo Open</th>
<th>ReSound Air</th>
<th>SeboTek</th>
<th>Vivatone</th>
</tr>
</thead>
<tbody>
<tr>
<td>Type of hearing aid</td>
<td>Digital</td>
<td>Digital</td>
<td>Digital</td>
<td>Digital</td>
</tr>
<tr>
<td>Channels, bands</td>
<td>20, 20</td>
<td>3, 3</td>
<td>4, 4</td>
<td>4, 12</td>
</tr>
<tr>
<td>Number of memories</td>
<td>3</td>
<td>1</td>
<td>4</td>
<td>4</td>
</tr>
<tr>
<td>Positions of signal processing unit</td>
<td>Behind-the-ear</td>
<td>Thin, hard tubing to transmit acoustic signal</td>
<td>Thin, hard wire housed in polyethylene tubing to conduct electric signal</td>
<td>Thin, hard wire supported by a moldable back bone wire to conduct electric signal</td>
</tr>
<tr>
<td>Sound delivering device in the ear canal and support feature</td>
<td>Fit n’ Go: soft plastic tip placed at the entrance of the ear canal and a retention hook to help holding the tubing in place</td>
<td>Thin plastic dome with holes placed at the entrance of the ear canal and a sports bend (plastic positioner) to help holding the plastic dome in place</td>
<td>Receiver module plus a soft mushroom tip placed in the bony portion of the ear canal</td>
<td>Receiver module placed at the entrance of the ear canal and a locking wire to help holding the receiver module in place</td>
</tr>
<tr>
<td>Modular?</td>
<td>Yes, the signal processing unit can be separated from the sound delivering device</td>
<td>Yes, the signal processing unit can be separated from the wire and the receiver module</td>
<td>No</td>
<td>No</td>
</tr>
<tr>
<td>Occlusion effect reduction mechanism</td>
<td>Open ear canal fitting</td>
<td>Open ear canal fitting</td>
<td>Deep canal fitting</td>
<td>Open ear canal fitting</td>
</tr>
<tr>
<td>Verifications</td>
<td>Verifications unavailable</td>
<td>The manufacturer reported -2.5 to -5 dB of occlusion effect compared to -2.5 dB of open ear occlusion effect (Christensen, 2004 AAS)</td>
<td>The manufacturer reported an average occlusion effect of -7.2 to -2.7 dB compared to -2.5 dB of open ear occlusion effect (Staab, 2003; Staab et al., 2004 AAS)</td>
<td>The manufacturer’s unpublished preliminary data showed less than 2 dB of insertion loss using real ear measurements</td>
</tr>
</tbody>
</table>

*These hearing aids are selected because they were commercially available at the time this paper was prepared.*
tivity in the low-frequency region, and they are subjected to the above-mentioned disadvantages associated with large vents.

Hearing aids with open-fitting sound-delivering devices also have limited high-frequency amplification. Several manufacturers use very thin tubing to connect between the receiver and the sound-delivery devices for cosmetic reasons. Yet, this thin tubing often limits the amount of high-frequency gain (i.e., 25 to 27 dB for Air according to Christensen, 2004, personal communications). This limits the fitting range at high frequencies to moderate-to-moderately severe hearing loss.

In addition, the amount of high-frequency amplification is limited by the effectiveness of the feedback reduction algorithm of the hearing aids. This is because the large vent provides minimum attenuation to sounds traveling from the receiver to the microphone and the primary attenuation is provided by the distance between the receiver and microphone. Therefore, for hearing aid candidates with more severe hearing loss, regular earmolds are recommended.

Another revisited approach used to reduce the occlusion effect is to reconfigure the components of behind-the-ear hearing aids. A conventional behind-the-ear hearing aid has a microphone, an analog or DSP unit, and a receiver housed in the case of hearing aid. The receiver output is coupled to the tubing and a custom earmold. The idea of reconfiguring the components of behind-the-ear hearing aids was explored many years ago. The very first reconfigured behind-the-ear hearing aid had a button receiver that received the processed sounds from the body of the hearing aid via a wire. The receiver was attached to an earmold, which delivered the processed sounds into the ear canal (Novak, 2004, personal communication).

SeboTek VoiceQ and Vivatone have recently launched newly designed behind-the-ear or postauricular canal (PAC, as SeboTek preferred) hearing aids that have receivers situated in the ear canal. The hearing aids from these two companies look very similar at first glance. Both hearing aids have a very small behind-the-ear hearing aid body, a moldable, hard and curved wire to transmit the electric signal from the hearing aid signal processor to the receiver, and a receiver placed in the hearing aid user’s ear canal. In addition, both companies use the modular design: the hearing aid body has a jack that receives a plug connected to the hard wire and receiver module (Figure 10).

These two hearing aids, however, differ in the intended receiver positions in the ear canal and the mechanisms used to reduce the occlusion effect. Specifically, the receiver/speaker of SeboTek VoiceQ is wrapped in a plastic covering and attached to a soft tip. The soft tip is designed to make full contact with the bony portion of the ear canal to deliver sounds close to the eardrum and to reduce the occlusion effect. On the other hand, the Vivatone receiver is held in the ear canal by a small locking wire. It is designed to be very small so that it is placed close to the entrance in the ear canal. The space surrounding the receiver lets the bone-conducted energy escape, thus reducing the occlusion effect.

The fitting range of VoiceQ extends to 80 dB HL at low frequencies and 90 dB HL at high frequencies, whereas the fitting range of Vivatone reaches 40 dB HL at low frequencies and 80 dB HL at high frequencies. The difference in their fitting range is mainly due to the differences in the distance of the receiver from the eardrum and the amount of low-frequency sounds escaping from the ear canal. Additional information on VoiceQ and Vivatone is listed in Table 2.

3.2. Verification and Limitations
The advantages of placing the receiver in the ear canal over the use of conventional earmolds are to eliminate the frequency response shaping and resonances associated with the tubing, and the reduction in sound pressure level as sounds travel from the receiver to the tip of the earmold.

SeboTek reported that the range of the occlusion effect for VoiceQ is between –2.7 to –7.4 dB when measured using the Occlusion Effect Meter (Staab, 2003; Staab et al., 2004). Staab (2003) reported that hearing aid users of SeboTek VoiceQ needed insertion training to place the receiver at the bony portion of the ear canal. If the receiver was placed in the cartilaginous portion, a higher occlusion effect was measured. Therefore, clinicians need to train hearing aid users for deep insertion to receive maximum benefit from SeboTek hearing aids.

Previous attempts to insert earmolds or canal portion of custom hearing aids into the bony portion of the ear canal have resulted in complaints of discomfort from hearing aid users. Staab (2004, personal communication) conducted a comfort test and the results indicated that 55 of
104 subjects ranked SeboTek VoiceQ with soft mushroom tips more comfortable than their own hearing aids during a test conducted in an office visit, 39 had no preference, and 10 found their own hearing aids more comfortable.

Vivatone's unpublished real-ear measurement data reported an insertion loss of less than 2 dB (Aronovici, 2004, personal communication). An unoccluded ear has an insertion loss of 0 dB. The limitations of Vivatone hearing aids are similar to the mini behind-the-ear Air and Perseo Open using the “Fit-and-Go” kit: a limited low- and high-frequency fitting range and the large-vent effect. Vivatone is suitable for people with nearly normal low-frequency hearing and a high-frequency hearing loss. The manufacturer also reported with 50-60 dB HL flat hearing loss were successfully fitted with Vivatone.

Figure 10. Two of the reconfigured behind-the-ear hearing aids. The receivers of (A) SeboTek VoiceQ and (B) Vivatone hearing aids is placed in the ear canal. VoiceQ utilizes the deep canal fitting mechanism whereas Vivatone uses the open ear canal fitting mechanism to reduce the occlusion effect. Figures provided by courtesy of SeboTek and Vivatone.
3.3. Strategy No. 3: Reducing the Vent Length

In the effort to reduce the occlusion effect, one hearing aid manufacturer has implemented stepped vents in its in-the-ear, in-the-canal and completely-in-the-canal hearing aids and Flex-Vent for their behind-the-ear hearing aids. Assume that the length of the hearing aid between the faceplate and the tip of the canal portion is 100%. Stepped vents have 15% tapered from the faceplate inward and 15% tapered from the tip of the canal portion outward. With this arrangement, the length of the stepped vents is reduced to 70% of that of a conventional vent (Christensen et al., 2004).

3.3.1. Verification

According to a study by Kiessling and colleagues, a completely-in-the-canal hearing aid with a stepped vent increased the occlusion effect for approximately 0.6, 1.8, and 2.5 dB at 200, 500, and 800 Hz, respectively, compared with a completely-in-canal hearing aid of the same brand with a traditional vent (Jenspersen, 2004, personal communication). Subjective ratings of the occlusion effect have also showed a slightly better averaged score for the traditional vent.

4. Challenge No. 4: Shell Manufacturing of Custom Hearing Aids

Traditionally, making custom hearing aids involves several procedures. First, the audiologist makes an impression of the ear canal and mails the impression to the hearing aid manufacturer. A cast is made to record the ear impression prior to any modifications; the shell technician examines, tapers, details, and trims the impression to the appropriate style; and the impression is dipped in hot wax. A second cast is made with a semitransparent jelly-like hydrocolloid, and liquid acrylic shell materials are poured into this cast. The liquid shell material is cured at room temperature or in an ultraviolet (UV) light chamber (Cortez et al., 2004).

After that, the liquid acrylic material is poured out and the lateral side of the shell is flattened so that the faceplate can be attached later. If a vent is desired, the shell technician drills a hole at the receiver end, runs a wire at the desired vent site, pours the liquid acrylic material, and cures the shell again. Finally, the technician drills another hole in the receiver end of the shell for the receiver tube, drills a hole on the faceplate for a vent, cuts the faceplate to fit the shell, puts the hearing aid components in the shell, glues the faceplate onto the shell, and buffs the shell (Cortez et al., 2004).

The traditional process of custom hearing aid manufacturing is labor intensive and is subject to mishandling and human error. For example:

1. The impression may be damaged or deformed in the mail.
2. It takes time to physically transfer the impression from the audiologist’s office to the manufacturing facility.
3. If any mistake is made in the process of trimming or handling, the damage is irreversible.
4. Uneven thickness may be generated during waxing, shell curing, and buffing. Thin areas can be easily damaged or cracked.
5. The components of the hearing aid and vents are fitted into the shell after the shell is made. Any miscalculation in the size of the shell can cost time and labor to remake the shell or delay the whole manufacturing process (Lesiecki, 2002; Cortez et al., 2004).
6. If the hearing aid is lost, the user has to go to the audiologist and start the whole process again. This causes customer inconvenience and hearing aid usage downtime.

4.1. Strategies: Laser Scanning and Shell Manufacturing Technologies

Several hearing aid manufacturers have recently adopted laser scanning and 3D printing technologies to manufacture custom hearing aid shells. The impression is scanned with laser to create 3D digital images of the impression. Then the shell technician finishes the digitized impression by detailing blemishes or imperfections in the computer images. The specifications of the functional structures of the custom shell are added to the image (e.g., thickness of shell wall, vent paths, and mounting frames for electronic components). The shell is then sent to be laser-printed (Cortez et al., 2004; Darkes and Voll, 2002; Lesiecki, 2002).

The laser printing process usually starts with slicing the digital image of the shell structure into thin cross-section layers by computer-aided design and manufacturing software. A series of digital files in the standard triangular language...
(STL) are created for these layers—each STL file contains information on one cross-section (layer) of the shell and different files correspond to different depth of the shell. The thickness of each layer is typically 0.1 to 0.25mm. The thinner the layers, the smoother the initial shell surface. A 3D printing machine uses these files to construct the shell layer by layer. Each layer is fused to the previous layer until a shell is formed.

Currently, hearing aid manufactures have adopted two 3D laser-printing processes: selective laser sintering and stereo lithography. The following section briefly describes these two processes, their advantages and limitations.

4.1.1. Selective Laser Sintering Printing Process
In the selective laser sintering process, a biocompatible nylon powder is used as the building material for hearing aid shells (Figure 11A, Lerner et al., 2002). This process has several steps:

1. The nylon powder is heated to just below its melting temperature and stored in one or two piston(s) on the side of the work surface.
2. The piston pushes the powder up and a roller spreads a thin, even layer of the powder onto the work surface.
3. The first layer's STL file guides a laser beam to heat the powder to the melting temperature at the location specified by the STL file.
4. The melted powder cools (sinters).
5. The work surface descends by one layer.
6. Steps 2 through 6 are repeated with subsequent STL files until the shells are completed (Amato, 2003).
7. As one batch of finished shells are suspended by the powder cake, another batch of shells can be printed by lowering the work surface to add space between batches and by repeating Steps 2 to 6.
8. The shells are removed from the work surface and the unsintered powder is removed.

4.1.2. Stereolithography Printing Process
In the stereolithographic process, a UV laser beam is used to harden photosensitive acrylic liquid resin layer by layer (Figure 11B, Lerner et al., 2002). This process can be explained by several simplified steps:

1. A platform is set with its surface just one layer below the level of photosensitive liquid resin.
2. The recoating bar spreads a thin, even layer of the liquid resin onto the platform.
3. The first layer's STL file guides a laser beam to harden the liquid resin at the locations specified by the STL files.
4. The platform descends by one layer.
5. Steps 2 through 5 are repeated with subsequent STL files until the shells are completed.
6. The shells are removed from the platform, and the external and internal surfaces are cured by ultraviolet light (postcuring).

As the shells are being cured in liquid resin, only one batch of shells can be made during each printing cycle.

Shells made by both processes have a matted finish. However, shells made by the stereolithographic printing process have a slightly smoother finish than those made by the selective laser sintering printing process, because powder has a more discontinuous nature than liquid (Cortez et al., 2004).

4.1.3 Advantages of Laser Shell Manufacturing
Manufacturing custom shells by using digital scanning and 3D laser printing technologies has multiple advantages. On the manufacturing side, these technologies reduce human errors in the traditional shell manufacturing process, increase productivity, and reduce labor. Any mistake made in the finishing and detailing process is reversible. The shell technician can simply retrieve the original image and restart the detailing process. In addition, the technician can visualize the placement of the internal components before the shell is printed. This reduces the chance of the receiver pointing toward the canal wall and thus reduces the chance of feedback. Moreover, being able to view the internal components allows for precise location of a vent tube or a measurement tube (for the probe microphone during real ear measurements) prior to shell fabrication. Further, each laser printing operation takes about 3 to 8 hours and produces a batch of 80 to 200 shells, depending on the size and style of the shells printed (Cortez, 2004, personal communication; Lesiecki, 2004, personal communication). This significantly increases the speed of the shell manufacturing process and reduces the labor involved.

On the clinical side, hearing aid manufacturers reported that shells created by instant manufacturing technologies have more accurate fit, more even thickness, less feedback, better comfort,
Figure 11. The illustrations of two three-dimensional laser printing processes used in manufacturing custom hearing aid shells. (A) In the selective laser sintering process, a laser beam is used to melt solid-state materials (nylon powder) to form custom shells. Several batches of shells can be printed in one printing process. (B) In the stereolithographic process, a laser beam is used to cure liquid state materials (acrylic resin) to form custom shells. Reprinted and modified with permission from Lerner et al (Aug. 2002), Foundry Management and Technology.
lighter shell weight, and higher durability than the conventional handmade shells (Cortez et al., 2004; Lesiecki, 2002; Fabry, 2002). Should the hearing aid need to be remade because it is lost or damaged, the audiologist can order the hearing aid without the hearing aid user making an office visit or the audiologist making a new impression (Cortez et al., 2004; Fabry, 2002). The manufacturer can reprint the custom shell from computer files.

The laser shell-manufacturing technologies have also paved the way for the making of a new generation of modular hearing aids. The core of a modular hearing aid is a compact module of pre-assembled electronic components. Shells made for modular hearing aids will have part of the faceplate with a hole to accommodate the compact module. Modular hearing aids will greatly simplify the custom hearing aid manufacturing process. Should a hearing aid malfunction, the compact module can be replaced in the audiologist’s office and the hearing aid user will experience minimal downtime for hearing aid use (Powers, 2004, personal communication).

According to one of the manufacturers, Siemens, the above laser shell-making technologies are in phase I of the shell making revolution. In phase II, health care professionals will be able to scan the impression in their offices, send the impression images to the manufacturer via the Internet, and order the hearing aid online. In phase III, the ear impression procedure will be totally eliminated. Clinicians will scan the image of the external ear by inserting a probe or a light stick in the ear for an e-impression (Lesiecki, 2002). Currently, some hearing aid dispensing offices are already equipped with the phase II ear impression scanning machines.

4.2. Verification and Limitations

Researchers at Phonak investigated the differences between conventional UV-cured and the laser-cured shells on several dimensions (Darkes and Voll, 2002). They reported that 43 experienced subjects readily accepted laser-cured custom shells. However, there were no long-term (12 weeks) differences between the UV-cured and laser-cured custom shells in ease of insertion, ease of removal, comfort at the time of insertion, comfort after 1 to 2 hours in the ear, comfort after 4 to 8 hours in the ear, irritation, retention, sound of their own voice, feedback, general appearance, and overall satisfaction. Nonetheless, 84% of the subjects chose to keep the laser-cured shells when they were given the option to keep one type of their hearing aids.

A study conducted by Widex reported that hearing aids made by the conventional UV-cured, laser nylon and laser acrylic resin shells, have similar in-situ thresholds, insertion gain for conversational speech, speech recognition at low input levels, subjective comfort, and potential occlusion (Cortez et al., 2004). Laser acrylic resin shells, however, were reported to have 5 to 6 dB and 2 to 3 dB more available gain for low-level inputs than the conventional shells and laser nylon shells, respectively, although the attributing factors were not explained.

Several cautions were recommended by the hearing aid manufacturers for handling the laser-printed shells. First, the nylon material has a melting point of 170°C. Should shell modification be needed, clinicians need to use a handheld drill or a standard drill with a lower speed and/or less pressure than would be used for the conventional acrylic shells (Cortez et al., 2004; Fabry, 2002). The laser acrylic shells have a melting temperature similar to that of the conventional acrylic shells, therefore, drill speed and pressure similar to that applied to the conventional acrylic shells can be used.

Second, acrylic shells are more brittle than nylon shells, which are less likely to break in a fall. In addition, Cortez and colleagues (2004) also recommended that a probe-vent for in-situ measurements be installed in laser shells during the manufacturing process. Laser shells have a better fit in the ear canal that may lead to the compression of the probe tube and the generation of inaccurate results. Inserting the probe microphone through the vent is not recommended, because it may result in the probe microphone pointing to the upper canal wall.

Currently, some manufacturers are looking into the digital light process, another 3D printing technology that has been modified from large-screen projection apparatus. The reason for the change is to overcome some problems created by the current 3D printing technologies. For example, the nylon shells have a lower melting point than the desired acrylic materials, and the laser-cured acrylic shells require more postcuring procedures. In addition, these shells may not be bio-compatible with some hearing aid users (Fabry, 2004 personal communication).

As technology advances, more sophisticated instant 3D scanning and manufacturing tech-
nologies will emerge. These technologies will continue to enhance the quality and quantity of shell manufacturing and reduce the time for making, repairing, or remaking of custom hearing aids.

5. Other Recent Developments

5.1. Improved Transducers
The design of hearing aid microphones and receivers has been improved in recent years. The size of these components has been reduced for the same output and performance so that they can fit into smaller hearing aids. On the other hand, if the size of the receiver is kept the same, a higher output is now possible. In addition, the new generation of microphones and receivers use internal damping or ferro-fluid damping to remove the peaks in their frequency responses. This allows the removal of damping screens and reduces the chance of wax or debris clogging the microphone or receiver port (LoPresti and Kochkin, 2003; Thompson, 2002). Additional efforts have also been made to redesign microphones to reduce internal noise and receivers to provide higher shock protection and higher output (Flynn & Flynn, 2004; Thompson, 2002).

5.2. Switchless, Programmable, and Active Telecoils
Automatic switching telecoils are now available for hearing aid users. They have a sensor to detect the electromagnetic field emitted by the telephone handset. When a handset is placed near the hearing aid, the hearing aid automatically switches to the telecoil mode. It can also automatically switch back to the microphone mode when the handset is moved away (Marshall, 2002). This automatic switch can reduce the problems associated with manual switching, such as delays in answering the phone and inability to switch because of reduced finger sensitivity or young age.

In addition, telecoils have also advanced in other areas:

1. Some telecoils have frequency responses similar to a typical hearing aid microphone. This feature is beneficial because it makes such telecoils acoustically transparent to the signal processors. The frequency responses of conventional telecoils are fixed. The frequency responses of new programmable telecoils can be altered to match the needs of the hearing aid user.
2. Active telecoils have filters to attenuate the interference created by digital cellular phones.
3. Most of the existing telecoils use the through-hole packaging. Some new telecoils use the surface-mount packaging, which is more durable and resistant to heat and shock (Marshall 2002).

5.3. Channel-Free Amplification
Bernafon has implemented a new form of compression system, channel-free compression, in its Symbio hearing aids. In a multichannel compression system, frequency- and level-dependent signal processing is accomplished by dividing the incoming signal into multiple frequency channels. In a channel-free compression system, frequency- and level-dependent signal processing is achieved by processing the incoming signal as a whole without dividing it into different frequency channels.

The mechanism of the channel-free compression can be explained in two stages: hearing aid fitting and real-world usage. During the hearing aid fitting process, gain settings of Symbio can be adjusted at nine audiometric frequencies at nine input levels (i.e., $9 \times 9 = 81$ points of adjustment) (see the part of figure with white dots in Figure 12). The nine audiometric frequencies are 250, 500, 750, 1000, 1500, 2000, 3000, 4000, and 6000 Hz and the nine input levels are 30, 40, 50, 60, 65, 70, 80, 90, and 100 dB SPL. Clinicians can provide level-dependent amplification to hearing aid users by adjusting the frequency response at each of the nine input levels (i.e., by altering the positions of the 81 points in the fitting software). Internally, Symbio’s signal processing algorithm transforms frequency responses of the nine input levels into frequency responses for nine loudness levels (i.e., from 10 to 90 sones) (Scheller, 2004, personal communication).

When the hearing aid is worn in the ear, Symbio’s signal processing algorithm calculates the centroid frequency of the incoming signal and estimates the normal loudness of the incoming signal from the calculated centroid frequency and
the overall input level. Then, the frequency response corresponding to the estimated loudness level and the centroid frequency is used to amplify the incoming signal. If the estimated loudness falls between two loudness levels (e.g., 58 sones), the frequency response is interpolated from the two immediate loudness levels (i.e., 50 and 60 sones). For loudness levels lower than 10 sones or higher than 90 sones, the signal processing algorithm extrapolates linearly based on the frequency responses at 10 and 90 sones, respectively. For very high input levels, the microphone output is limited to avoid digital clipping, which is similar to peak clipping but occurs in the digital domain (Scheller, 2004, personal communication). Spectral smearing occurs when the relative spectral content of sounds are altered. Researchers have reported negative effects of spectral smearing on speech intelligibility (Boothroyd et al., 1996; ter Keurs et al., 1992, 1993; van Schijndel et al., 2001).

In a study conducted by Dillon and colleagues (2003), the performance of Symbio was compared with the performance of three other digital hearing aids, Phonak Claro 211 dAZ, Widex Senso Diva, and ReSound Canta 770D. Subjects with normal hearing and hearing loss listened to and rated the processed signals from these four hearing aids in six listening conditions: male discourse, female discourse, male discourse in an impulse noise, piano music, own voice, and quiet room with no sound. Paired comparisons were

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Figure 12. The sequence of channel-free amplification. When a signal is picked up by the microphone: (1) The hearing aid signal detection and analysis unit estimates the centroid frequency of the incoming signal for the instance (Time 1 or Time 2); (2) The loudness of the incoming signal is estimated based on the centroid frequency and the overall sound pressure level; (3) The incoming signal is amplified using the frequency response corresponding to the loudness level. (The part of figure with white dots is provided by courtesy of T. Scheller at Bernafon).
conducted with advanced signal processing algorithms (e.g., directional microphone, feedback reduction algorithms, noise reduction algorithms) turned off, except that noise reduction algorithms were activated at the moderate level in the male-discourse-in-noise and the quiet-room conditions.

Dillon and colleagues reported no significant differences among the hearing aids in most of the listening conditions for both groups of subjects, except that normal hearing subjects preferred Symbio in the quiet-room condition (Dillon et al., 2003). The same study was also reported in the National Acoustics Laboratories 2002-2003 Annual Report. Keidser and colleagues (2003) reported that normal hearing subjects preferred Symbio and Triano to Claro and Diva when listening to male discourse in impulse noise and subjects with hearing loss preferred Symbio to Claro when listening to piano music.

5.4. Open-Platform Hearing Aids

Traditionally, DSP chips in hearing aids are “hard-wired” for a particular set of signal processing algorithms that are set before the hearing aids leave the manufacturing facility. The first open-platform digital hearing aid, ReSound 5000, was launched in 1998, and several hearing aid manufacturers shared the platform. “Open platform” means that the manufacturer or clinicians can choose the software to be downloaded into the hearing aid. In this first generation of open-platform hearing aids, different manufacturers can download their signal processing algorithms to the same DSP chip to make different brands of hearing aids.

In 2004, GNReSound launched the second generation of open-platform digital hearing aids, Canta Open. With Canta Open, clinicians can decide if they want to download the Canta or Air fitting algorithm for the particular hearing aid user. The signal processing algorithm in Canta Open is changed by simply choosing the name of the fitting algorithm in the fitting software.

According to GNReSound, this feature adds flexibility and convenience for clinicians in the hearing aid fitting process. If one fitting algorithm is not optimal, clinicians can try another one without switching to a different hearing aid. The availability of open platform hearing aids also reduces the number of hearing aids need to be stocked in clinicians’ office for demonstration purposes.

5.5. Binaural Hearing Aids

Siemens has recently launched Acuris, the first generation of binaural hearing aids. They and their remote control unit (if any) communicate via short-range radio wave which shifts between 114 and 120 kHz. Acuris has 11 transmission Identity Codes. Code 1 is the default when all Acuris and their remote controls are shipped out of the manufacturing facility. When two Acuris hearing aids are fit binaurally, they analyze sounds in the environment, automatically share information, and switch modes and programs simultaneously when a change in the environment is detected. A change made in the settings by the remote control or the buttons on either hearing aid is simultaneously made in the other hearing aid (Powers and Stenfanic, 2004, personal communication).

According to Siemens, the ability of a pair of binaural hearing aids to communicate in real time can help older users with less sensitive fingertips gain better control of the hearing aids. Conventionally, the volume control wheel and the program change button are implemented on the same hearing aid, but older users may have difficulty differentiating the wheel from the button. If two Acuris are fit binaurally, the program change button can be put on one hearing aid and the volume control wheel on the other. When a user changes the volume or the program of one hearing aid, the same change is automatically made in the opposite hearing aid (Powers and Stenfanic, 2004, personal communication).

All Acuris remote controls and hearing aids are defaulted to Identity Code 1 when they are shipped out of the manufacturing facility. If two people are expected to use Acuris in close range, the clinician can program their remote controls and hearing aids to a different identity code to avoid cross-control or interference (Powers & Stenfanic, 2004, personal communication).

To enhance the experience of binaural hearing aids, hearing aid users need to be aware of two facts. First, the communication among the two hearing aids and the remote control can be affected if they are within 30–40 inches of cathode ray tube (CRT) computer monitors or television screens. Second, whenever a hearing aid is connected to direct audio input, it ceases to receive program-change commands from the remote control or the other hearing aid. In addition,
it always defaults to Program I whenever the input is disconnected from the direct audio input. If the user connects a direct audio input monaurally and switches the program on the other hearing aid, he/she needs to turn off and then turn on the hearing aids to re-establish telecommunication. This is because only if both hearing aids are in the same program (e.g., both on Program I), they can communicate with each other and their actions are synchronized (Powers & Stenfanić, 2004, personal communication).

5.6. Rechargeable Hearing Aids

Magnatone has marketed in-the-ear digital and analog programmable rechargeable hearing aids. The hearing aids are put in a charging case to recharge the battery. These rechargeable hearing aids free people with dexterity difficulties from the need to change batteries. Some models are also compatible with Zinc Air batteries. The manufacturer claims that every 8 hours of charging time allows 16 hours of use. The rechargeable battery lasts up to 5 years. Other advantages of rechargeable hearing aids include drying the hearing aids while charging and protecting the environment from discarded batteries. Verification data on the performance of these hearing aids compared to other hearing aids are not available.

5.7. Behind-the-Ear Frequency Modulation/Booth FM Receivers

FM receivers, once bulky and inconvenient, can now be attached to or incased in behind-the-ear hearing aids because of the advances in the miniaturization of electronic components. Several manufacturers have marketed behind-the-ear hearing aids with built-in FM receivers or ear-level FM receivers that can be attached to behind-the-ear hearing aids. Hearing aids with such built-in or ear-level FM receivers are cosmetically appealing and lightweight. One caution with these miniaturized FM receivers is that they drain the batteries much quicker than a hearing aid alone; therefore, it is important for the clinician to consult with the manufacturer for estimated use time and instruct the client to check the battery frequently.

5.8. Wireless Bluetooth FM System

Phonak has recently launched the SmartLink system, which can serve as a remote control of hearing aids, a hand-held directional microphone system, an audio signal receiver and transmitter, an FM system, and a two-way Bluetooth cellular phone inter-agent. Specifically, the SmartLink is a fully functional remote control for hearing aids and FM systems. It is also a hand-held directional microphone unit with three microphone modes: omni-directional, directional, and super-directional. In addition, SmartLink can act as the microphone and the transmitter of an FM system and send the signal to hearing aids. The transmission range is roughly 10 meters and it can be extended to roughly 30 meters using the external lavalier cord.

Another use of SmartLink is to accept electric signals from audio sources (e.g., microphone, CD player) and send the signals via the FM transmitter to both of the hearing aids (in case of binaural fitting). It also has a digital synchronizer that can detect other FM signals within the transmission range when the hearing aid is turned on. The digital synchronizer automatically change the transmission frequency of the transmitter to an unused frequency, and synchronize the reception frequency of the receiver. If an interfering signal occurs after the hearing aid is turned on, the user can manually change the transmission frequency using the remote control. The digital synchronizer then automatically changes the frequency of the FM receiver.

Further, SmartLink allows hands-free cellular phone operation. It can receive signals from Bluetooth cellular phones and transmit the signals to the hearing aids so that the user can hear the cellular phone signals in both ears. It can also pick up the hearing aid user’s voice and transmit it back to the cellular phone. The maximum distance between SmartLink and the cellular phone is approximately 10 meters. Other features of SmartLink include a fine-scale noise canceler that is implemented in Phonak Perseo hearing aids and a digital voice-enhancement algorithm which supposes to enhance speech (http://www.phonak.com/professional/products/fm/smartlink.htm).

5.9. Bluetooth Wireless Programming Interface

Programmable and digital hearing aids are conventionally programmed by using computers through HIPRO interface boxes or through manufacturer’s special programming devices. A new NoahLink programming interface using the Bluetooth wireless technology became available
in January 2003. The NoahLink is a portable device that can be worn in front of a hearing aid user. It provides a wireless two-way communication between the computer and the hearing aid(s) (i.e., the wired connection between the computer and the HIPRO box is eliminated). NoahLink allows users to walk around or outside the clinic to test the effectiveness of hearing aid programs. If further changes need to be made in the hearing aid program, there is no wire to reconnect (Florian, 2003). In addition, effective programming can be conducted within 5–6 meter radius of the Bluetooth adaptor without the “line-of-sight” requirement. The programming signal can be transmitted through glasses or walls. The data transmission rate, in practice, is roughly 12 times faster than the HIPRO box (Brown, 2003).

Several cautions need to be exerted when using the NoahLink. First, the transmission range is significantly reduced if the Bluetooth adaptor is placed on the floor behind the computer. The performance can be improved by raising the Bluetooth adaptor to the level of table top. Second, although the manufacturer of NoahLink, HIMSA, endorses the use of Bluetooth adaptors from several companies, the adaptors may vary in their transmission range and data transmission rate. Third, the data transmission rate reduces as the distance between the NoahLink and the Bluetooth adaptor increases. The data transmission rate becomes essentially the same as the HIPRO box when this distance is increased to 6 meters. Fourth, if the USB version is used, direct connection between the computer and the Bluetooth adaptor is recommended. Connecting the Bluetooth adaptor to a USB hub can significantly compromise the performance (Brown, 2003).

6. Applications of Hearing Aid Technologies to Other Devices

The recent breakthroughs in hearing aid technologies have inspired engineers and researchers to apply advanced hearing aid technologies to other amplification and communication devices. For example, Chung and colleagues (2004a, 2004b) have explored the feasibility of applying advanced hearing aid signal-processing algorithms to enhance cochlear implant performance. They have applied directional microphones and noise reduction algorithms as preprocessors to cochlear implants and reported improvements in listening comfort, overall preferences, and speech understanding of cochlear implant users.

It is conceivable that other advanced hearing aid technologies (e.g., advanced microphone technologies, noise reduction algorithm, and automatic switches) can also be used to enhance the performance of cochlear implants. Chung and colleagues are also exploring the benefits of using hearing aid preprocessors with electric output (Figure 13) in everyday life environments. Preliminary data are promising.

Hearing aid technologies have also been applied to create hearing protectors. Walker’s Game Ear Incorporated markets hearing protectors that look very similar to hearing aids. In addition, Chung and colleagues have modified high-performance digital hearing aids and created smart hearing protectors that can provide speech-dependent and environment-dependent attenuation.

SoundID have also applied multichannel wide dynamic range compression and noise reduction algorithms to cellular phones to enhance speech understanding and perceived sound quality of cellular phone users.

7. Conclusions

It is truly a very exciting time in the hearing health care profession. New technology breakthroughs happen almost daily. With the continuous shrinkage of the size of computer chip, the increase in processing speed, and the reduction in power consumption, it is expected that more automatic and sophisticated options will be available in digital hearing aids. In addition, researchers and engineers are applying hearing aid technologies to other amplification and communication devices. New devices that take advantage of the combination of technologies are also emerging.

This paper can only review a fraction of the technologic advancements and by no means should be treated as all-inclusive. In addition, the signal processing algorithms implemented in a hearing aid are designed to work together and the output of one algorithm may alter the decision-making process of other algorithms. For example,
the adaptive directional microphone, the noise reduction algorithm and the compression system of Oticon Syncro interact with each other. This paper reviewed individual algorithms but not the synergy or interaction among them.

In the midst of the technology explosion, the following thoughts should be kept in mind during the hearing aid selection and fitting process:

1. Different manufacturers may have different names for similar signal processing functions. For example, adaptive feedback-reduction algorithms in hearing aids are called Digital Feedback Suppression, Dynamic Feedback Cancellation and Diva Active Feedback Cancellation. On the other hand, some algorithms may have similar names yet they may differ in their signal-processing strategies and thus have different effects on the processed signals. For example, the feedback-reduction algorithms implemented in Phonak Valeo and Oticon Syncro sound very similar, Digital Feedback Canceller for Valeo and Dynamic Feedback Canceller for Syncro. However, Valeo uses the notch filtering strategy whereas Syncro uses the phase cancellation strategy. Clinicians need to understand the underlying mechanisms, the operational advantages, and the limitations of the technologies used in hearing aids.

2. Although previous research has not shown a significant increase in the satisfaction of hearing aid users (Wong et al., 2003), high-performance digital hearing aids do offer many more signal processing options. They are worth a try for difficult listening environments and for challenging fitting cases.

3. Clinicians need to understand a hearing aid user’s unique listening needs and life style to choose the hearing aids that can offer the most appropriate features. In addition, clinicians

![Figure 13](image-url). One of the implementations of a hearing aid preprocessor for a cochlear implant. A cochlear implant user is wearing an in-the-ear hearing aid preprocessor in conjunction with his cochlear implant. The hearing aid preprocessor is used to preprocess sounds. Its electric output is fed into the direct audio input of his cochlear implant. The hearing aid preprocessor is provided by courtesy of Oticon, Demark.

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need to project realistic expectations, explain the technologies, and inform the user of how to get the maximum benefits from the chosen technology.

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