Acoustic Feedback and Other Audible Artifacts in Hearing Aids

Jeremy Agnew, Ph.D.

Director of Product Development, Starkey Laboratories, Inc.
1110 Elkton Drive, Unit E, Colorado Springs, CO 80907

Acoustic feedback has been described as "whistling," "howling," "screeching," "screaming," "squealing," "whining," "ringing," "humming," "buzzing," "oscillating" and by various other names. The high-pitched whistling of a hearing aid experiencing acoustic feedback is an irritating sound for the hearing aid wearer and for nearby individuals. Suppressing these irritating squealing noises is not easy. Thus dealing with acoustic feedback is still a prevalent problem that plagues clinicians and wearers alike. Though specific figures are considered by manufacturers to be proprietary information, industry experts estimate that as many as 10% to 15% of in-the-ear hearing aid products are likely to be returned to the factory within the first 90 days after manufacture for feedback-related problems. Obviously this adds to the overall cost of hearing aids to the dispenser, and anything that can be done to help reduce these returns will ultimately benefit the wearer.

This issue is intended to provide comprehensive information on the origins and characteristics of acoustic feedback in hearing aids and to discuss its minimization or prevention. Although it will primarily discuss acoustic feedback, for completeness the discussion will also include the audible symptoms of electrical and electromagnetic feedback and pickup. These also occur in hearing aids and are frequently confused by the wearer with acoustic feedback. Some discussion of the symptoms of both may help the clinician to understand and resolve complaints of oscillations and other audible artifacts in a troublesome hearing aid fitting. It is useful for the clinician to be able to distinguish between various manifestations of oscillation and other audible sounds in order to be able to logically identify the problem and counsel the wearer appropriately.

Before starting a detailed discussion of feedback, an important point should be made concerning terminology. Acoustic feedback in a hearing aid fitting produces a form of instability and the resulting audible oscillation. It is caused by a sound wave from the output leaking back to the input. Though all acoustic feedback of the correct phase and magnitude produces an undesired form of oscillation in a hearing aid, not all oscillation is due to acoustic feedback. In precise terms the objectionable audible sound produced by a hearing aid due to acoustic feedback should be called *audible oscillation due to acoustic feedback*. Through common usage, this more accurate term has generally been abbreviated simply to *acoustic feedback*, though in reality acoustic feedback is the cause of the problem and not the audible effect. However, to comply with common usage the term *acoustic feedback* will be used consistently throughout the text to refer to the unpleasant and undesired squealing and screeching that occurs in a hearing aid and which is caused by the leakage of amplified output sound back to the microphone.

A second point should also be made about terminology. For simplicity, hearing aids will be discussed in two general categories, unless the text specifically addresses a certain type of hearing aid. The term *in-the-ear hearing aid* will be used to generically refer to ITE (in-the-ear), ITC (in-the-canal) and CIC (completely-in-the-canal) hearing aids, unless a specific one of the three types is stated and discussed. This general term is intended to contrast to *BTE (behind-the-ear) hearing aids*. This distinction is made because there are two significant differences in construction between the generic categories of in-the-ear and behind-the-ear hearing aids. These differences affect the causes of feedback, the symptoms, and possible solutions. Also, note the use of lower-case letters for *in-the-ear* to denote the generic connotation. The upper-case initials *ITE* will be used to specifically refer to the full-concha "In-The-Ear" hearing aid. Italicized type will be used in various places in the text to draw attention to certain important concepts and to terminology which is not typical for audiological use.

The first major difference in construction between custom in-the-ear hearing aids and BTE hearing aids is the obvious difference between the two types in the methods of housing the electronic amplifiers and the transducers, such as microphones, receiver and telecoils. BTE hearing aids are generally manufactured on a production line and have identical internal construction for a specific model of hearing aid. Frequently, BTE hearing aids have cavities in the case for the microphone and receiver in order to provide acoustic isolation to minimize internal acoustic feedback. Custom in-the-ear hearing aids, on the other hand, are individually handcrafted at final assembly, though most of the faceplate circuit subassemblies are made on a production line with standardized layouts. This means that transducers for in-the-ear hearing aids are fitted into the shell wherever space allows. The close proximity of transducers makes a custom in-the-ear hearing aid more prone to feedback than a BTE hearing aid, unless considerable care is taken during construction. The effects of these differences will be described later in this issue.

The second significant difference in construction is that BTE hearing aids use an earhook, acoustic tubing and an external earmold to provide the acoustic coupling from the hearing aid to the ear. In-the-ear hearing aids, on the other hand, are built to fit directly into the ear. Thus, while generic comments can be made about the two type of hearing aids concerning items such as venting and canal length, there may be some differences in approaches to feedback control which will need to be described in more detail. Finally, there will be instances in which it is important to specifically discuss differences between in-the-ear products (i.e. ITE versus ITC versus CIC), and these instances will be made clear in the text.

BASIC PRINCIPLES OF ACOUSTIC FEEDBACK

In this age of ubiquitous electronic amplification, it is often observed that if a speaker using a public address system in a conference room or an auditorium stands too close to the loudspeaker, then a loud and obnoxious squeal may occur. This is acoustic feedback. A portion of the sound coming from the loudspeaker has been picked up by the microphone, has been amplified, and then radiated back into the room. This situation is shown diagrammatically in Figure 1. Part of this amplified signal is picked up by the microphone, is re-amplified, and is subsequently re-radiated into the room where it is again picked up by the microphone, and so forth. This repeating cycle of sound amplification, radiation and pickup continues until the system is no longer stable and oscillation occurs. The audible manifestation of this instability is a loud and overwhelming squeal. This sound is obnoxious and is irritating to both the speaker and the audience.

Thus acoustic feedback is a circle of amplification, where amplified sound is continuously re-amplified to the point at which a tonal squeal occurs. The specific tonality of the squeal is determined by the electronic characteristics of the amplifier combined with the acoustic characteristics of the microphone, the room and the loudspeaker. Due to the varied dimensions, and the reflection and absorption characteristics of different structures, different rooms produce squeals with different tonal characteristics.

Though the obvious and most common manifestation of acoustic feedback is a squeal, feedback in itself is sometimes desirable. Electronic feedback may be intentionally created in a circuit in order to achieve desired results. For example, electronic feedback may be used to create tones for use in test equipment, such as audiometers. To accomplish this, a signal is deliberately fed back around an amplifier in a controlled fashion to create an electrical tone similar to that resulting from acoustic feedback. Changes in the values of the components in the electronic circuit are used to

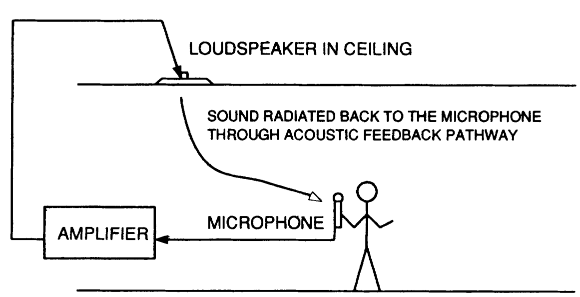

Figure 1. Example of acoustic feedback in a public address system.

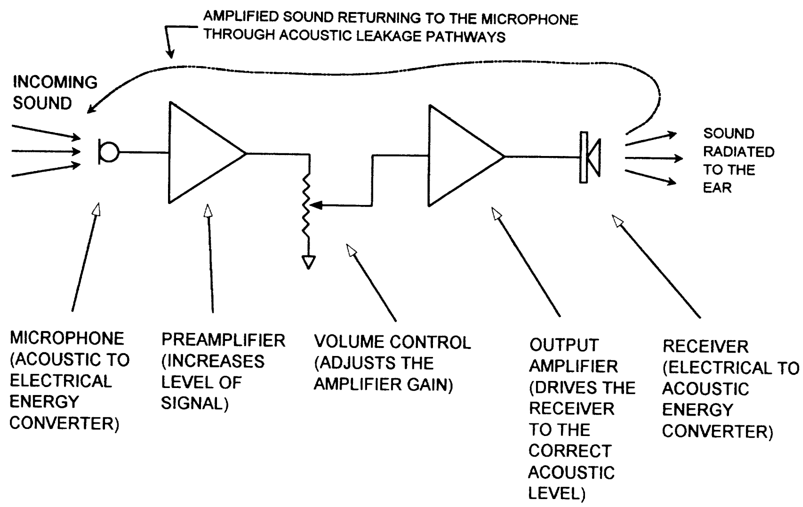

Figure 2. Airborne pathway from a hearing aid leading to acoustic feedback.

change the frequency of the tone, in order to create a range of test frequencies.

Though the example of acoustic feedback at the beginning of this section was applied to a public address system, the same principle applies on a smaller scale to a hearing aid. Amplified sound transmitted to the ear canal from the receiver is radiated out through the vent, or via various other pathways (such as acoustic leakage between the earmold or hearing aid shell and the wall of the ear canal via a pathway called *slit-leak*), back to the microphone. Then it is amplified and re-radiated out of the ear canal, where it is picked up again by the microphone, re-amplified and so forth. Figure 2 shows a schematic representation of the acoustic feedback pathway in an ITE hearing aid that can lead to acoustic feedback. Figure 3 illustrates an ITE hearing aid in place in the ear, showing potential acoustic leakage pathways.

Though slit-leak and an adequate seal to the ear are very important, acoustic feedback caused by slit-leak and venting may or may not be present. The problem created by this leakage also depends on the amount of gain provided by the hearing aid. If the gain is quite low there may not be enough sound radiated out through leakage pathways to cause acoustic feedback. Thus, the effect of any venting in this instance will only be to reduce low frequency amplification.

The probability of acoustic feedback is greater in a hearing aid than with a public address system because the microphone and receiver in a hearing aid are in fixed locations very close to each other. Also, it is generally not possible to move the microphone further away from the receiver to prevent feedback, as may be done with a public address system. There are exceptions to this rule in specialized fittings where the microphone may

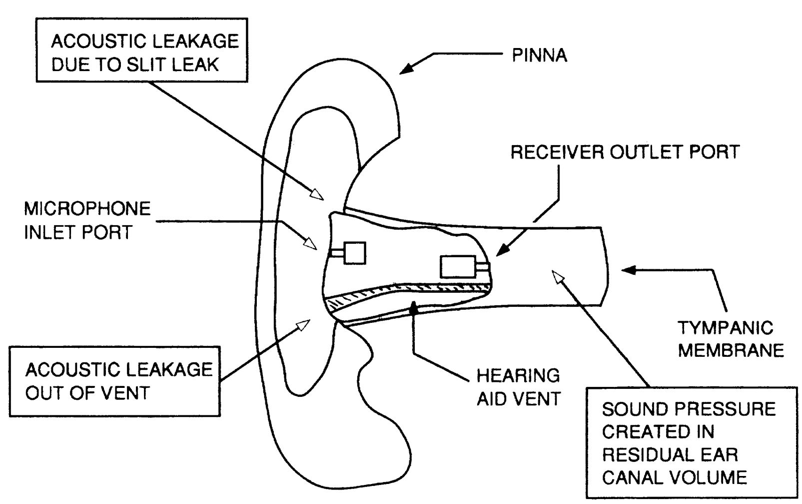

Figure 3. ITE hearing aid placed in the ear, showing potential acoustic feedback pathways through the vent and through slit leakage.

sometimes be separated from the receiver. This solution will be described later.

In summary, it can be seen from this fundamental description of feedback that several conditions have to exist for acoustic feedback to occur and be sustained in a hearing aid:

1. Some of the sound radiated from the receiver has to leak out of the ear canal and be picked up by the microphone,
2. Amplification has to occur,
3. The amplified sound has to be re-radiated from the receiver and ear canal back to the microphone.

Thus far, this description of acoustic feedback has been very basic and qualitative to ensure that the reader has a fundamental understanding of the mechanism which results in the familiar squeal heard from a hearing aid.

From this generalized description, it would appear that acoustic feedback could occur at any frequency. However, as is well-known by hearing aid wearers and clinicians, feedback usually occurs at a frequency which gives the audible acoustic screech a distinctly tonal quality. Most wearers also empirically note that the pitch of feedback may be altered by changing the acoustic conditions surrounding the hearing aid. For example, moving a cupped hand nearer to or further away from a hearing aid usually changes the pitch of the audible squealing sound. Thus, though the theoretical potential exists for feedback to occur at any frequency, in reality, it only occurs at one or two frequencies. The reason for this is determined by the acoustics of the feedback environment and relates primarily to the phase of the signal passing through the hearing aid circuit.

THE IMPORTANCE OF PHASE IN THE CREATION OF ACOUSTIC FEEDBACK

Sound being returned to the microphone from the receiver has a certain *amplitude* and *phase*. The amplitude of a signal, such as a sine wave, is simply the magnitude of the signal. The term *phase*, however, has two meanings in relation to feedback. The first is the *phase angle* of a sine wave, which represents the progression of a wave through one cycle, or 360°.

Consider the 1000 Hz sine wave displayed in Figure 4, which shows one full cycle of the wave and part of another. The graph shows that one full cycle (360°) of the wave occurs in the first 1 millisecond, then an incomplete cycle continues out of the right side of the graph. The figure is plotted as amplitude versus time, and shows the correspondence of time and frequency on the x-axis. The amplitude of this sine wave starts at zero, rises to a positive maximum value after 0.25 milliseconds or 90°, returns to zero at 0.5 milliseconds or 180°,

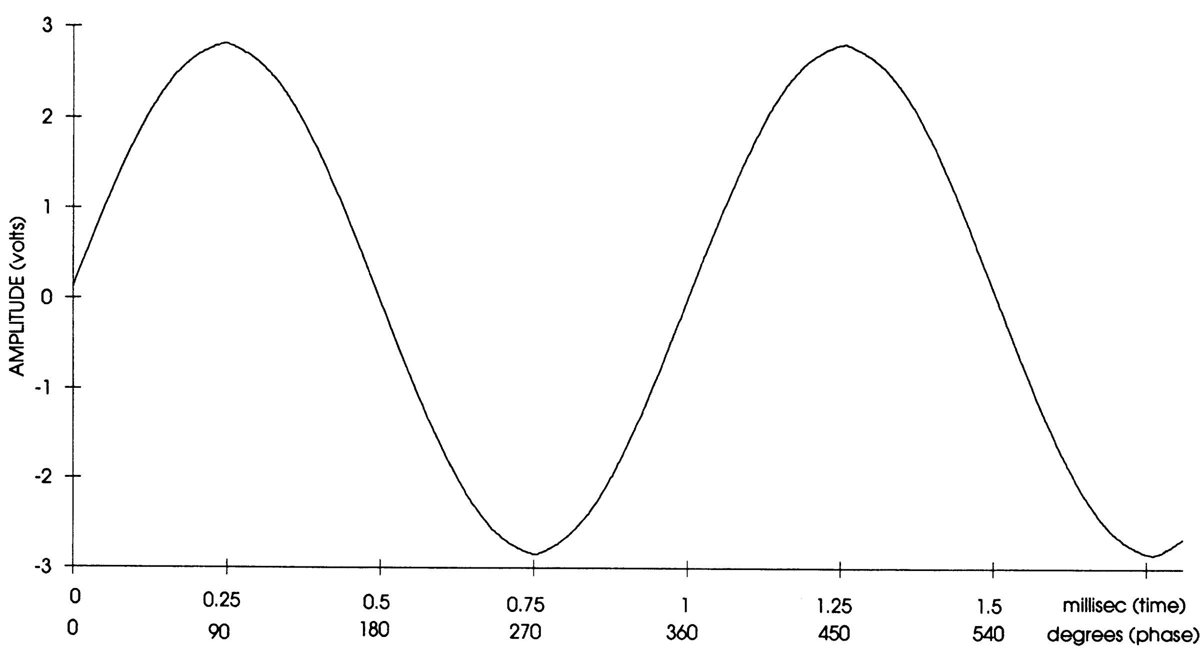

Figure 4. Plot of amplitude (y-axis) of a 1000 Hz sine wave versus time (x-axis, upper scale), also shown is phase angle in degrees as an alternate scale (x-axis, lower scale).

reaches a negative maximum at 0.75 milliseconds or 270°, and returns to zero at 1 millisecond or 360°. At 360° the amplitude of the wave is the same as it was at 0° and the sine wave starts through another cycle. At 450° the wave has repeated to be the same maximum positive value that it had at 90°, and so forth. The time for a sine wave to repeat (the period) is calculated by remembering that time is the inverse of frequency, where frequency is specified as Hertz and defined as the number of cycles per second of the sine wave. In this case, the period of the sine wave would be calculated as: 1 second/1000 cycles = 0.001 seconds = 1 millisecond. Thus 1 millisecond coincides with 360° for a 1000 Hz sine wave.

A sine wave completes a full cycle (one period) with a phase angle of 360°, as shown in Figure 4. When one cycle of the wave is completed, another starts, and the wave repeats itself. Thus the amplitude of a constant sine wave will be the same at each multiple of 360° of phase angle along the wave. Conversely, it can be seen that a sine wave burst of N cycles will complete $N \times 360°$ of phase angle. These mathematical details are not particularly important for the qualitative description of acoustic feedback in this issue, and hence can be accepted or ignored, as desired.

Figure 5 is an illustration of the frequency response and phase angle versus frequency for a typical hearing aid. Figure 5a illustrates the frequency response curve; Figure 5b shows the corresponding phase angle at each frequency. The measurement was made in a sound box and is only intended to introduce the concept of phase angle as it relates to hearing aids, therefore the specifics of the measurement conditions are not important. By examining Figures 5a and 5b at any desired frequency, the gain and corresponding phase angle of the sound traveling through the hearing aid can be identified.

Phase is plotted in Figure 5b such that 0° is at the center of the graph, with 0° to +180° in the upper half of the graph and 0° to −180° in the lower half of the graph. Because of the way it is dis-

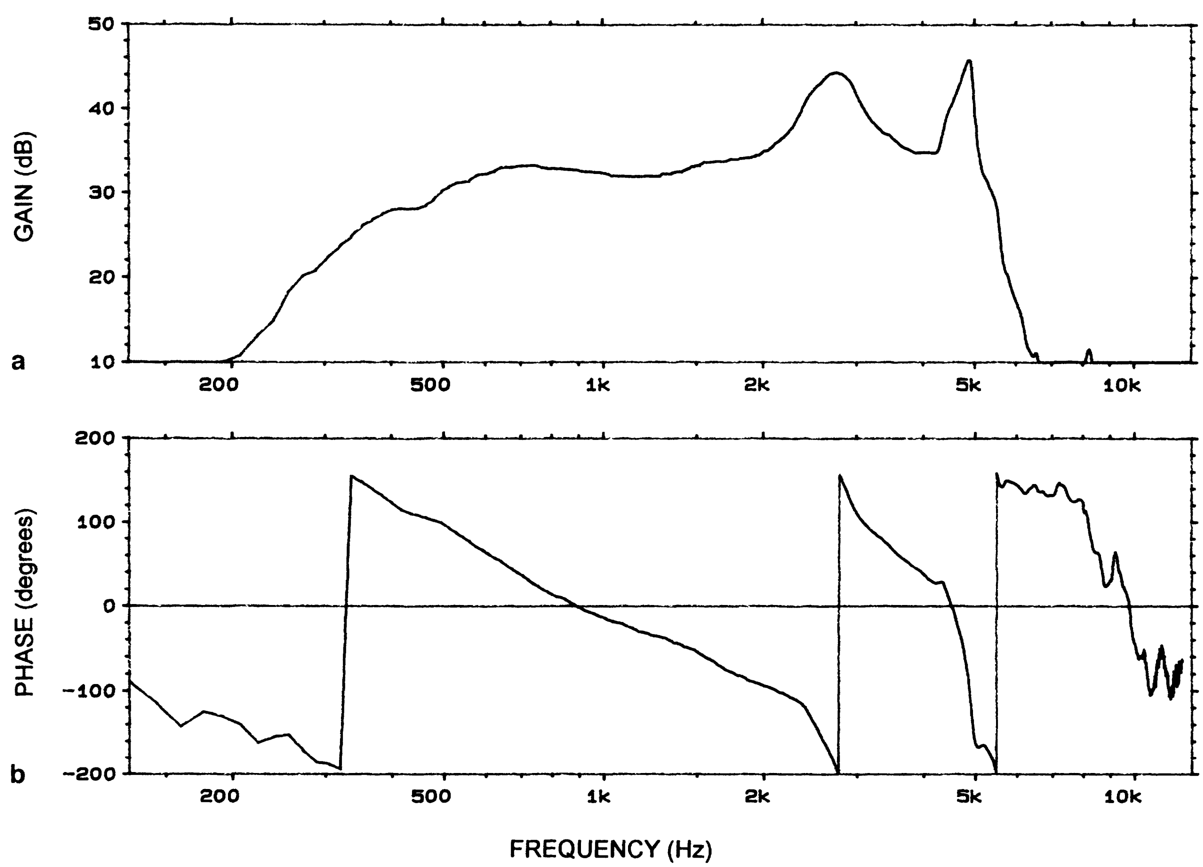

Figure 5. a: Plot of gain versus frequency in a sound box for a typical hearing aid. b: Plot of phase versus frequency for the same hearing aid under the same conditions.

played, the phase curve may appear to be a little confusing. Each time the phase reaches $-180°$, the equipment used to measure phase reverses it back to $+180°$ (since 180° plus 180° equals 360°, or a complete cycle); thus, the reading is the same at both $+180°$ and $-180°$. In Figure 5b, phase inversions occur at 320 Hz, 2768 Hz and 5456 Hz. These inversions make the graph appear as a series of saw teeth, with vertical lines appearing on the graph where phase changes between $-180°$ and $+180°$. This is referred to as *wrapping up* the phase and is used to simplify the display into a smaller graph. If the phase graph were to be *unwrapped*, that is displayed with an additional 360° of y-axis for every cycle, it would rapidly drop off the bottom of the page and the scale on the left axis would have to be labeled 0°, $-360°$, $-720°$, $-1080°$, and so forth.

The second meaning of phase in the context of feedback is *phase difference*, which is measured as the time difference between the maximum amplitude points of two sine waves. Note, however, that phase difference does not necessarily have to be referenced to the maximum point on a sine wave. It is also possible to use the minimum point, or any other easily-referenced amplitude point on the wave, as long as it occurs at the same relative position on both waves. For example, it is not correct to use an amplitude point on the descending quadrant of one wave and refer it to the same amplitude point on the ascending quadrant of the other wave.

The concept of phase difference can be understood by considering Figures 6a, 6b and 6c. Figure 6a illustrates two sine waves that have 0° phase difference between the top and bottom parts of the figure, and thus are considered to be *in-phase*. The maximum amplitude points of the two waves coincide on the time scale. If these signals were added together, the resultant amplitude would be two times as large as either one presented as a single source.

Figure 6b shows two sine waves that have 180° phase difference, and thus are considered to be *out-of-phase*. Note that the maximum and minimum amplitude points of the waves are exactly opposite from one another on the time scale. When one sine wave is at maximum amplitude, the other sine wave is at a minimum. If these two signals were added, they would cancel each other out and the resultant signal amplitude would be zero. Thus, if these were two sound waves, there would be no resulting audible sound.

Figure 6c shows two sine waves that have an

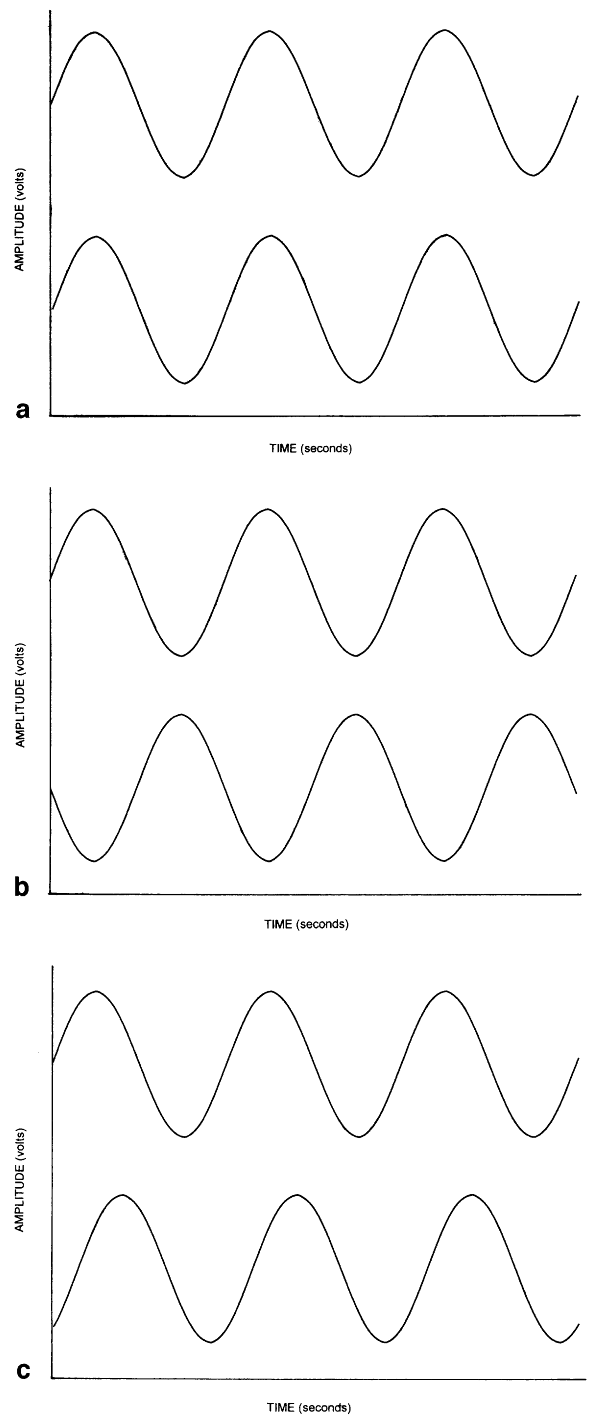

Figure 6. a: Two sine waves that are in-phase. b: Two sine waves that are 180 out-of-phase. c: Two sine waves with a 54 degree phase difference, showing the two waves between the in-phase and out-of-phase conditions.

arbitrary phase difference between 0° and 360°. The phase difference can be found from the time difference between the maximum amplitude points of the waves and, in this case, is 54°. The resulting wave, if both were added together, would be some combination of the two individual waves added together instant by instant.

The simple qualitative explanation to remember for understanding acoustic feedback is that if a leakage sound is fed back in-phase with the sound being amplified, the sounds reinforce each other and become larger in amplitude, eventually resulting in audible oscillations. This is called *positive feedback*. If the sound leaks back *out-of-phase* with the sound being amplified, the two sounds will partially or completely cancel each other and the resulting sound will be reduced or canceled. This is called *negative feedback*.

The significance of the type of information shown in Figure 5 for analyzing and understanding feedback is the occurrence of the in-phase conditions, as shown on the phase graph in Figure 5b. This is the condition in which the measured phase crosses the 0° line. At frequencies where the phase (Figure 5b) crosses the 0° line and the open loop gain (Figure 5a) is greater than 1, oscillation will occur. In Figure 5b, 0° phase crossings occur at 880 Hz, 4528 Hz, and 9792 Hz.

Since Figure 5 is provided for illustration purposes to introduce the concept of phase, these measurements were not of a feedback pathway, but were made of the forward path of a hearing aid in a sound box. Thus this data does not illustrate a potential feedback condition. To analyze a specific feedback situation on a wearer, the measurements shown in Figure 5 would be made of the open-loop feedback situation in the wearer's ear. This is because the specific environment surrounding the fitting must be included in the measurement, including the individual pinna, ear canal, fit of the hearing aid in the ear, and all the other individual wearer characteristics that would affect the feedback pathway. An illustration of such a measurement will be discussed and shown in Figure 10. Just as it is known that the frequency response curve will be different in the wearer's ear than when measured in a sound box, the phase will also be influenced by the specific circumstances of the fitting. Unfortunately, though measurement of phase could potentially be a valuable fitting tool, the difficulty of making the measurement accurately and repeatably on a wearer has so far precluded its use in clinical settings.

Before leaving this section, it should also be mentioned that the information in Figure 5 was obtained with a 2 cc coupler. Measurements in the real ear will show frequencies that could potentially result in oscillation to be different than those measured in a 2cc coupler, because the acoustic load on the hearing aid in the ear canal is different than the acoustic load of the 2cc coupler. The smaller residual volume of the real ear will alter the frequency response, typically increasing the overall and high frequency gains.

Figure 6a is important for understanding acoustic feedback because the signals add perfectly together in-phase. If one signal is the amplified acoustic signal and the other is the feedback signal due to acoustic leakage (which, in a real feedback situation, will occur at a reduced amplitude than that shown in Figure 6a), the amplified signal and the feedback signal will add. Thus, as the leakage signal is fed back and amplified several times, the two signals will continue to add together until eventually the signal amplitude is so great that oscillation occurs, which produces a characteristic audible squeal. By contrast, if the situation of Figure 6b occurs, in which the signal leaking back to the microphone is perfectly out-of-phase, the two waves cancel and subtract from each other. The resulting signal amplitude at the output is diminished. Thus acoustic feedback will not occur.

Attempts have been made to use the properties of phase to cancel feedback. One scheme that has been proposed is to automatically feed back part of the signal from the receiver to the microphone to create an acoustic negative-feedback loop to cancel potential acoustic feedback (Lichowsky, 1973). This strategy uses a special one-piece combination microphone and receiver with a thumbscrew adjustment. The adjustment controls the amount of acoustic leakage through a communicating passage between the microphone and receiver diaphragms. Thus far, such a transducer has not been readily available, and it is doubtful that it could conveniently and effectively be incorporated into an ITE or ITC hearing aid without considerable further research and development effort.

Another type of acoustic phase cancellation strategy has been proposed separately by Bordewijk (1991), Weinrich (1991), and Krokstad et al (1994). In these inventions, the feedback signal is detected by a second microphone which is placed inside the hearing aid and close to the receiver. The resulting signal from the second microphone is added out-of-phase to the primary signal, and thus cancels out acoustic feedback. Thus far use and testing of this strategy has been limited because the additional space required by the second microphone is often not available in

ITE or ITC hearing aids. If subminiature solid-state microphones based on semiconductor technology become a reality, further development of this type of strategy may receive more attention.

For readers interested in further mathematical details, the stability of feedback systems has been analyzed in detail by Nyquist (1932), who developed a theorem called the *Nyquist Stability Criterion*. "Stability" in this sense means the lack of oscillation in a circuit, system or hearing aid due to feedback. The essence of the theory is that a feedback system will become unstable (i.e. will have oscillation due to feedback) at any frequency at which the open loop gain is greater than 1 and the phase is 0° (or any multiple of 360° which, as shown in Figure 4, is equivalent to 0°). The system will remain stable if the open loop gain at these frequencies is less than 1. In a hearing aid, acoustic instability results in the familiar whistling or howling of acoustic feedback.

For those interested in a more rigorous analysis of the occurrence of feedback in hearing aids, Egolf (1982) and Egolf et al (1985) have presented a mathematical model of feedback, including discussion of the Nyquist Stability Criterion.

CHARACTERISTICS OF ACOUSTIC FEEDBACK

For the purposes of this text, acoustic feedback can be categorized into three general types, as shown in Figure 7:

1. *External acoustic feedback:* caused by some factor related to the fitting and not due to malfunction within the hearing aid. Examples of circumstances which may cause external acoustic feedback include very high gain, a vent with a large diameter or uncontrolled acoustic slit-leakage. This is the most common type of acoustic feedback.
2. *Internal acoustic feedback:* caused by leakage and subsequent audible oscillation within the hearing aid because of internal malfunction.
3. *Miscellaneous audible non-feedback sounds:* caused by factors interacting with the hearing aid which may lead to acoustic manifestations. This is not acoustic feedback, but is often mistakenly confused with it.

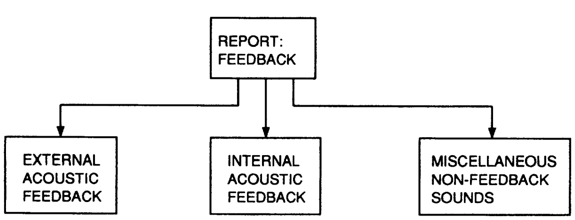

Figure 7. Categorization of reports of "feedback".

External acoustic feedback is caused by the leakage of sound from the receiver back to the microphone. It is caused by an interaction of a number of acoustical and mechanical factors which are inter-related in the fitting of a hearing aid. The presence of feedback can be caused by something as simple as the earmold or shell being seated improperly in the ear, which can result in undesired sound leakage and thus acoustic feedback. The impact of leakage on causing acoustic feedback is related to a number of factors. These factors include:

1. Residual volume of the ear canal. Decreasing the residual volume of the ear canal will raise the sound pressure in the remaining cavity and can initiate feedback. As a rough rule of thumb, every time the residual volume is reduced by one-half, the sound pressure increases by 6 dB.
2. Type of hearing aid. The smaller models of hearing aid, such as an ITC or CIC, place the microphone and receiver close to each other internally due to their small case size and also produce a shorter pathway between their external ports. This proximity produces a higher susceptibility to feedback.
3. Presence and configuration of venting. The larger the diameter of the vent, the more sound leaks out of the canal back to the microphone.
4. Amount of slit leakage. The larger the slit leakage present, the easier sound leaks out of the canal back to the microphone.
5. Fit of the hearing aid in the ear. The looser the fit in the ear, the more slit leakage will be present.
6. Length and diameter of the canal area of the shell or earmold. The longer the length of the canal portion of the hearing aid, the better the hearing aid seals to the ear and reduces the potential for feedback. The smaller the diameter of the canal portion with respect to the ear canal, the looser the fit and the more slit leakage will be present.
7. Wearer's pinna size and shape. The larger the pinna and the more it bends back towards the head, the more sound from vent-

ing or slit leakage is liable to be reflected back to the microphone.

8. Gain and frequency response of the hearing aid. The higher the overall gain, particularly the high frequency gain, the more prone a hearing aid is to have feedback.
9. Orientation of the hearing aid or earmold in the ear. If the receiver tube of the hearing aid or sound outlet bore of an earmold points towards the wall of the ear canal instead of at the eardrum, sound can more easily be reflected out of the ear and cause feedback.
10. Eardrum impedance. A stiffer eardrum is more likely to cause feedback than a more compliant one, due to more efficient reflection from the surface of the membrane.
11. The setting of the gain control. The higher the gain control setting, the more likely the possibility of feedback.

All these factors influence the occurrence of acoustic feedback and the frequency at which it occurs. Coughing, chewing, sneezing, yawning, talking, tilting the head, bringing a hand up to the face, use of the telephone, the proximity of reflective surfaces and placing a hat on the head can also initiate feedback in a hearing aid which borders on having an unstable feedback environment.

Because of these variables, acoustic feedback can be a very elusive phenomenon. It can occur at different frequencies with the same hearing aid at different times and under different acoustic conditions. The audible pitch of the feedback may vary smoothly as acoustic conditions change, or it may jump between different frequencies. Feedback may even occur at more than one frequency at the same time.

Acoustic feedback typically occurs at or close to the high frequency peaks in the hearing aid frequency response since these peaks occur at the frequency of the greatest gain. The peaks are primarily a result of inherent electromechanical receiver resonances that affect the acoustic output of the receiver. These peaks will vary with the dimensions of the specific model of receiver used in the hearing aid. The peaks are particularly influenced by the mechanical case size of the receiver, since different case sizes result in differing electromechanical resonances. Feedback tends to occur at one primary frequency and, due to saturation of the amplifier, this primary frequency will produce audible multi-frequency tones which are harmonics of the primary frequency (Agnew, 1993).

Feedback typically occurs in the region of high frequencies between 2000 Hz and 5000 Hz. It is often initiated by the large amount of high frequency gain typically used to successfully fit high frequency sensorineural hearing loss. It should be noted that a hearing aid that has moderate gain and a relatively flat frequency response is less likely to cause feedback than a hearing aid with greater gain and an emphasized high frequency response. Since acoustic feedback is initiated by excessive high frequency gain, the higher the overall gain, the more likely feedback is to occur. Specifically, the higher the gain in the high frequencies, the more likely feedback is to occur.

When discussing acoustic feedback with wearers, it may be helpful to counsel them that there are certain situations in which feedback may be inevitable. These include improper insertion of an ITE hearing aid or BTE earmold into the ear; turning the gain control to maximum setting; or cupping a hand around the hearing aid. Further discussion of these points can be found in Mynders (1982).

Stages of Feedback

Though acoustic feedback is often perceived as a single strident acoustic squeal, the initiation of feedback consists of several stages. These various stages, from initiation to a full-blown squeal, are shown in Figure 8a, 8b and 8c. Figure 8a shows the frequency response of a hearing aid with the gain control turned to a setting slightly below that at which feedback would occur. This was the typical operating frequency response of the hearing aid. There was an additional peak present around 500 Hz which is not usually seen in responses measured in the 2cc coupler. This was created by resonance of a slit leak in the acoustic coupling, which was deliberately included to simulate acoustic leakage when fitted in the ear.

As the gain control was increased slightly, the additional amplification caused a greater amount of sound to leak out of the vent and back to the microphone. As a result, the frequency response, as shown in Figure 8b, began to reveal a sharp peak at the potential feedback frequency which, in this case, was around 1800 Hz. The occurrence of this "spike" in the frequency response curve is called *sub-oscillatory feedback*, to indicate that the hearing aid is just below the point of delivering continuous audible oscillation. Cox (1982) has discussed the effects of sub-oscillatory feedback on the frequency response of vented hearing aids.

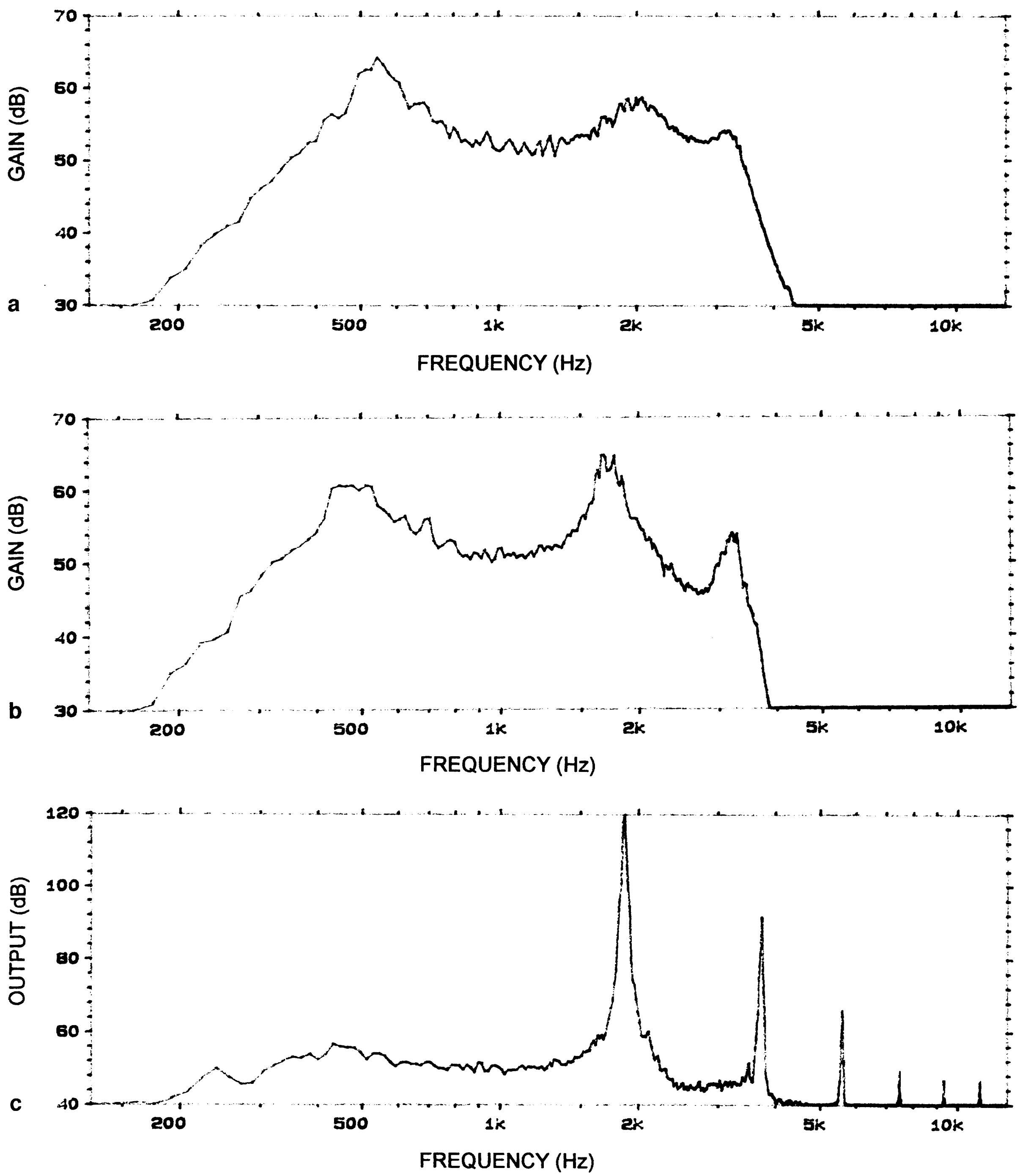

Figure 8. Frequency response curves from a hearing aid showing the various stages of feedback. a: No acoustic feedback; the gain of the hearing aid is set to the level below that at which feedback occurs. b: Sub-oscillatory feedback; note spike at around 1800 Hz which will develop into feedback. c: Acoustic feedback occurring at 1872 Hz; also seen are harmonics of this frequency.

The sharp peak of sub-oscillatory feedback that occurred around 1800 Hz in Figure 8b will also result in poor sound quality (Preves and Newton, 1989). When the hearing aid is in this condition, transient loud sounds often promote random ringing and whistling noises in the acoustic output. This is the result of intermittent stimulation of the hearing aid in and out of feedback by sound inputs with high energy at this frequency. Whether the ringing becomes audible at times or not, the presence of these sharp peaks in the frequency response creates a very objectionable-sounding form of distortion, particularly in the 2000 Hz to 3500 Hz region (Langford-Smith, 1960).

At this point, the feedback in this hearing aid was not continuous or audible, though a wearer might have complained of intermittent feedback, intermittent ringing noises, or intermittent reverberation. If these complaints are reported by a wearer, one of the best ways to test for the presence of sub-oscillatory feedback is to use the headphones often provided with probe tube measuring equipment to listen to what the wearer is hearing while the hearing aid is functioning in the ear. If the ringing or intermittent whistling ceases when the gain control is decreased slightly, then the gain control was probably set just to the point of sub-oscillatory feedback. This is a highly unstable state that usually progresses immediately into sustained audible oscillations.

Unfortunately the clinician is often left in a quandary for a solution if this problem occurs. It may be necessary to use a high gain control setting in order to restore audibility and speech intelligibility to the wearer; however, if the gain control is set to this position, sub-oscillatory feedback may occur and produce intermittent ringing. A compromise may have to be made, which could include the use of a damper, the reduction of a high frequency potentiometer setting, or the use of acoustic filtering. These options will be discussed later.

As the gain control setting of the hearing aid was increased, the resulting increased gain caused a sustained oscillation at a single frequency. This oscillation created the output graph shown in Figure 8c. This figure indicates a primary feedback frequency at 1872 Hz and the harmonics of 1872 Hz. What happened was that the sub-oscillatory feedback frequency was amplified to the maximum amount that the hearing aid amplifier, receiver and battery could provide. At this point, the amplifier went into saturation, the output signal became a square wave and distortion was created. This distortion is revealed in Figure 8c as the primary feedback frequency (1872 Hz) and a series of harmonics of this frequency appearing across the entire spectrum. In reality, the transition from the situation shown in Figure 8a to that of Figure 8c will be essentially instantaneous and the transition state in Figure 8b is not commonly observed. In Figure 8c note the change of scale on the y-axis. This is because hearing aid saturation is driving the hearing aid receiver to deliver its maximum SSPL90 of 120 dB.

Measurement of Open Loop Gain and Phase

It is important to measure the open loop gain of the feedback leakage shown in Figure 2 to fully understand the feedback. What is measured is the leakage portion of the feedback, since that is the portion of the loop causing the problem. It is useful to generate graphs as in Figure 6a and 6b of the leakage pathway, in order to be able to predict the open loop gain and phase conditions of feedback under a specific acoustic or fitting situation. Unfortunately, as mentioned earlier, making this measurement on a wearer is difficult in practice. This technique is primarily used in research settings rather than in clinical practice.

The other problem related to the measurement of phase is the need to develop a procedure to adjust the hearing aid following the measurement. In other words, though it may be academically interesting to measure the open loop gain and phase associated with feedback, it is also necessary to provide a clinical method to use this information to reduce its occurrence. Currently research is under way at several institutions to try to bring some variation of this technique into the clinic for more widespread use, and hopefully such a system will appear in the future. A greater importance will be placed on phase information when digital signal processing (DSP) hearing aids with the ability to readily manipulate phase are available. Meanwhile, in anticipation of such developments, it is important to at least understand the techniques and measurements involved.

Since the acoustic conditions associated with a hearing aid fitting will change the occurrence and characteristics of feedback, it is necessary to perform this open loop measurement with the hearing aid in the ear, as it will be worn. By making these measurements under the conditions of the actual fitting it is possible to accurately measure the open loop gain and phase around the feedback loop. The equipment necessary for making these measurement is shown schematically in Figure 9. This diagram is very similar to Figure 2, but Figure 9 contains the addition of the test equipment which is necessary to make the measurements. Note that a resistor has been substituted for the volume control for this measurement.

Acoustic feedback is produced by a constantly-energized closed loop, with the leakage signal to

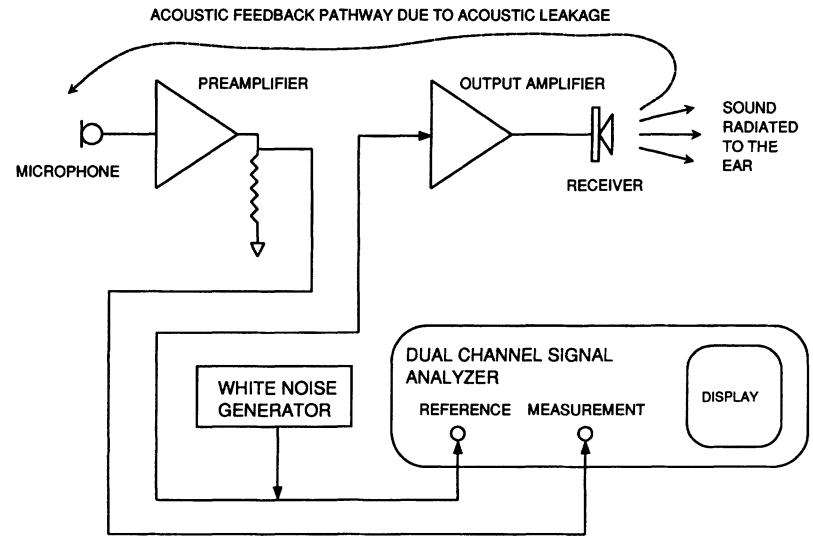

Figure 9. Block diagram of the technique for measuring the open-loop gain and phase response of a hearing aid using a dual-channel signal analyzer.

the input being continuously re-amplified. If the loop is closed and the hearing aid is feeding back, it is not possible to make accurate gain and phase measurements since the hearing aid is saturated by the oscillation signal. Thus, the other difference between Figure 2 and Figure 9 is that the feedback loop through the amplifier and acoustic pathway has been opened where the test equipment has been added. Opening the loop stops the audible oscillation and allows the open-loop gain and phase measurements to be made.

These measurements require the use of a dual-channel signal analyzer. Again this makes the technique is more readily suited for a research laboratory than a clinician's office. However, it is important to understand the principle of the measurement because this description reveals the basis for understanding feedback. In addition, this principle is the basis for the adaptive cancellation filter strategy currently used in some hearing aids which will be described later.

To make the measurement, the amplifier portion of the feedback loop is opened at a convenient place in the circuit, shown in Figure 9 as being between the output of the preamplifier and the input of the output amplifier. Breaking the loop stops the audible feedback, but keeps the external acoustic leakage part of the feedback loop intact to make the measurement.

A signal generator is used to inject a random white noise signal into the output amplifier and receiver. The direction of the signal flow shown in Figure 9 is revealed by the direction of the arrows. Thus, the white noise test signal comes out of the signal generator, goes into the output amplifier and receiver and, at the same time, goes into the reference channel of the dual channel signal analyzer. The measurement channel of the analyzer receives the leakage signal from the output of the preamplifier of the hearing aid that has been fed back through the air and into the microphone. Since this is a two-channel measurement, the reference channel of the signal analyzer is attached to the output of the signal generator to provide a reference signal for comparison to the measurement channel.

The principle of the measurement is:

1. The noise generator creates a sound in the ear canal through the output stage and receiver of the hearing aid. This signal is monitored by the reference channel of the analyzer.
2. This sound leaks from the receiver through the undesired acoustic feedback pathway.
3. This leakage sound is picked up by the hearing aid microphone.
4. The feedback signal due to leakage is measured by the analyzer after amplification by the preamplifier.

Thus, the measurement includes the acoustic feedback pathway just as it would be during operation of the hearing aid in the ear. The results of a measurement of a hearing aid under these conditions is shown in Figure 10, which displays both open-loop gain and phase measured in a 2 cc coupler. The amplitude of the sound leaking back to the microphone through the vent is shown in Fig-

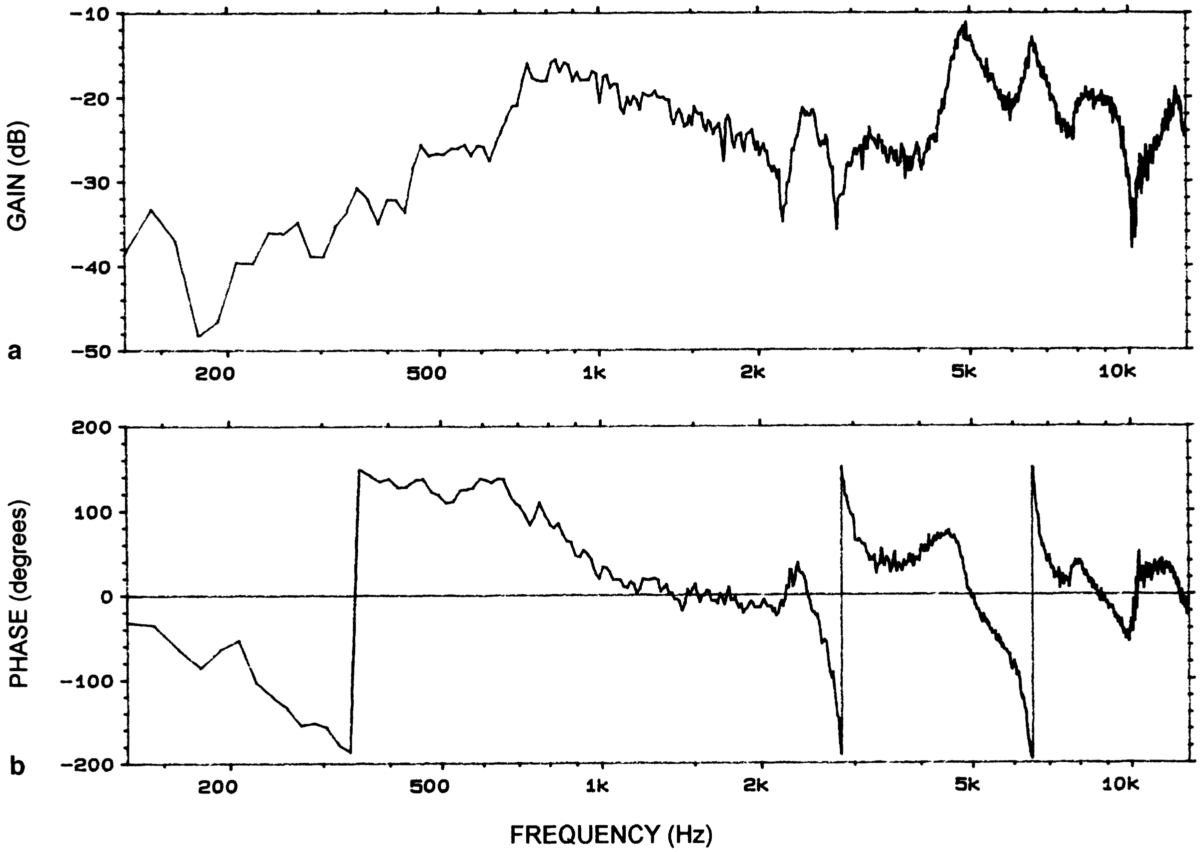

Figure 10. Open loop gain and phase measured by the equipment shown in Figure 9.

ure 10a. In this particular case the gain control of the hearing aid was turned down to avoid feedback.

The phase of the leakage signal is shown in Figure 10b. The critical zero crossings occur at approximately 320 Hz, 2800 Hz and 6300 Hz. However, since the gain is substantially less than 1 at each zero crossing (as well as across the entire frequency range), this hearing aid environment was stable and no feedback could occur.

Gain Available Prior to Feedback Oscillation

Two considerations determine the amount of gain available in a hearing aid before feedback. One is to avoid sub-oscillatory feedback; the other is to obtain the maximum gain that is necessary to produce an effective fitting.

Skinner (1988) has suggested that insertion gain should be 4 dB to 8 dB less than values at which audible feedback occurs to avoid the effects of intermittent sub-oscillatory spikes on the sound delivered to the wearer. Bisgaard and Dyrlund (1991a) have suggested that a gain of 5 dB below that at which feedback first occurs, may be necessary to ensure a stable system.

The maximum amount of insertion gain available will depend on the type of fitting. For example, hearing aids fitted with IROS (Ipsilateral Routing Of Signal) earmolds may achieve only 30 dB of gain when coupled to large ear canals. When the ear canal is normal or small then the gain can often be increased to up to 40 dB (Courtois and Berland, 1972). Kates (1988) has stated that, in most instances, venting a BTE earmold or ITE shell will limit the maximum insertion gain to about 40 dB. In addition, the maximum gain will be less if larger vents are used. A study by Gatehouse (1989) showed that a BTE hearing aid with a forward-facing microphone could achieve a maximum gain of 33 dB to 40 dB using a 2 mm vent. He reported that these values will vary with frequency.

Earmolds or cases with very small vents (2.6 mm to 4.0 mm) have been shown by Kates (1988) to increase the maximum possible stable insertion gain to between 60 dB and 80 dB. A BTE using a non-vented, well-fitting earmold should achieve maximum gains of approximately 60 dB (Grover and Martin, 1974). Skinner (1988) has claimed that "it is difficult, if not impossible, to prevent feedback with any earmold when gains of more than 60 to 65 dB are used. p. 271" Expanded discussions of this topic are given in Kuk (1994) and Valente et al (1996).

It should be pointed out that the above figures were based on values for linear hearing aids, and the use of non-linear hearing aids may lead to quite different results. For example, when compression is activated in some hearing aids, phase relationships through the amplifier change drastically, thus substantially altering the post-compression feedback characteristics. Another significant factor is the frequency response of the hearing aid. A wide-band hearing aid with high frequency emphasis is more likely to experience feedback than one with restricted high frequency gain.

Feedback When Using a Telephone

Acoustic feedback while using the telephone is a problem for hearing aid wearers. The problem is particularly difficult when using a hearing aid or earmold containing venting. Placing the telephone receiver near the ear forms a reflective surface that directs sound from the vent or from slit leak back to the microphone and can immediately initiate feedback. Reducing the gain control setting to prevent acoustic feedback is not always satisfactory, since this also reduces audibility and thus defeats the intended use of the hearing aid. Solutions to allow acoustic use of the telephone without incurring feedback are generally not particularly satisfactory.

Possible solutions for using the telephone without incurring feedback may include:

1. Use of a telephone coil. This eliminates the microphone as a source of acoustic feedback.
2. Use of direct audio input (DAI), with the appropriate adapter system for the telephone. This also eliminates the microphone as a source of acoustic feedback.
3. Use of a programmable hearing aid that reduces the bandwidth of the hearing aid to match the telephone frequency response and reduce the high frequency gain (Agnew, 1991).
4. Use of circuitry that automatically alters the phase of the signal through the hearing aid to prevent violation of the Nyquist Stability Criterion when the telephone is brought close to the ear.
5. Use of various self-adhesive foam rings or snap-on plastic extension devices that fasten over the telephone receiver to block ambient noise and acoustically seal the telephone to the ear (Grimes and Mueller, 1991).

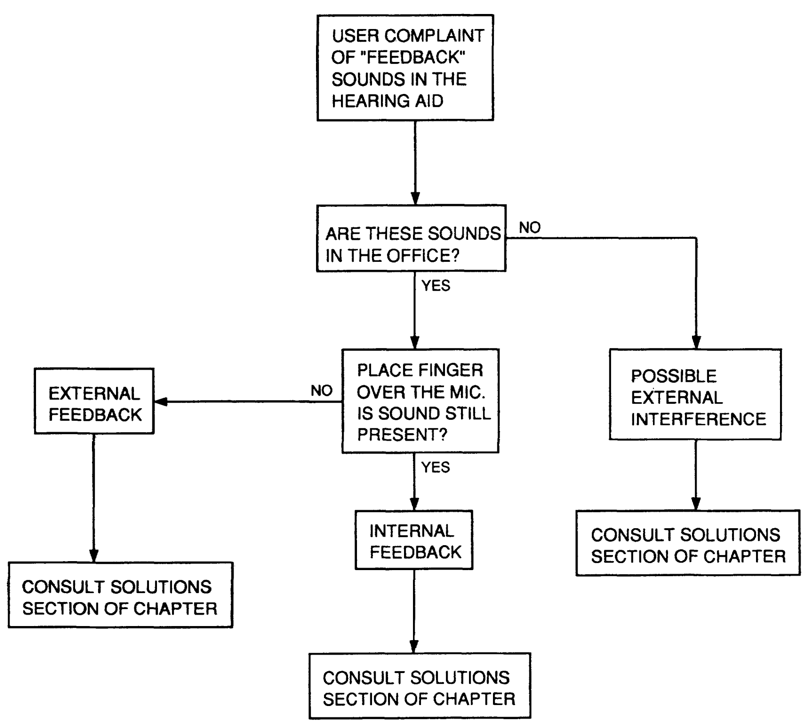

Figure 11. Differentiation of causes of "feedback sounds" in a hearing aid.

CAUSES OF EXTERNAL ACOUSTIC FEEDBACK

The most common problems with feedback are those related to the hearing aid fitting. Figure 11 outlines an easy method to differentiate between the three types of undesired sound (external and internal feedback, and external interference) that may be produced by a hearing aid.

Problems related to external acoustic feedback can be divided into two basic categories:

1. *External acoustic feedback that can be resolved by the clinician and still maintain the prescribed fitting.* This category comprises a multitude of problems which require minor adjustment of the hearing aid by the clinician during the fitting process. These modifications can be performed and still maintain the integrity of the prescribed fitting. These problems can often be resolved by judicious modification of the initial fitting. Examples of solutions are the reduction of high frequency gain, the addition of a damper, or the reduction of vent size.
2. *External acoustic feedback that cannot be resolved by the clinician and still maintain the prescribed fitting.* These comprise fittings in which the prescribed fitting cannot be maintained because it defies the laws of acoustics. In these situations, modifications will not resolve the problem if the goal is to maintain the desired acoustic characteristics of the fitting. An example of this is fitting a high gain hearing aid with a large-diameter vent and high frequency emphasis. It is possible that this fitting cannot be achieved without feedback because of the required gain and the leaking acoustic feedback pathway through the vent. The only solution to prevent acoustic feedback in this case may be to alter some characteristic of the fitting. Either the overall gain or the high frequency gain may have to be decreased, or the vent diameter reduced. Thus some aspect of the initial fitting will have to be compromised to avoid acoustic feedback, and the clinician will need to decide which provides the best compromise.

In most cases, external acoustic feedback results from one or more of the causes outlined in Figure 12. These include:

A. **Acoustic Leakage:**
1. Inaccurate ear impressions.
2. Poor fitting of the shell or the earmold.
3. Tubing not properly sealed to an earmold.
4. A crack in the earhook, tubing or earmold.

B. **Hearing Aid Characteristics:**
1. Excessive high frequency gain.
2. Proximity of components due to hearing aid case style.
3. Large vent diameter.

C. **User Characteristics:**
1. Sound reflection due to the shape and size of the pinna.
2. Exceptionally high ear canal resonance.
3. The presence of excess cerumen in the canal.
4. Orientation of the canal tip towards the wall of the ear canal.
5. Mandibular motion causing slit leak.

D. **Miscellaneous:**
1. The presence of nearby reflective surfaces.

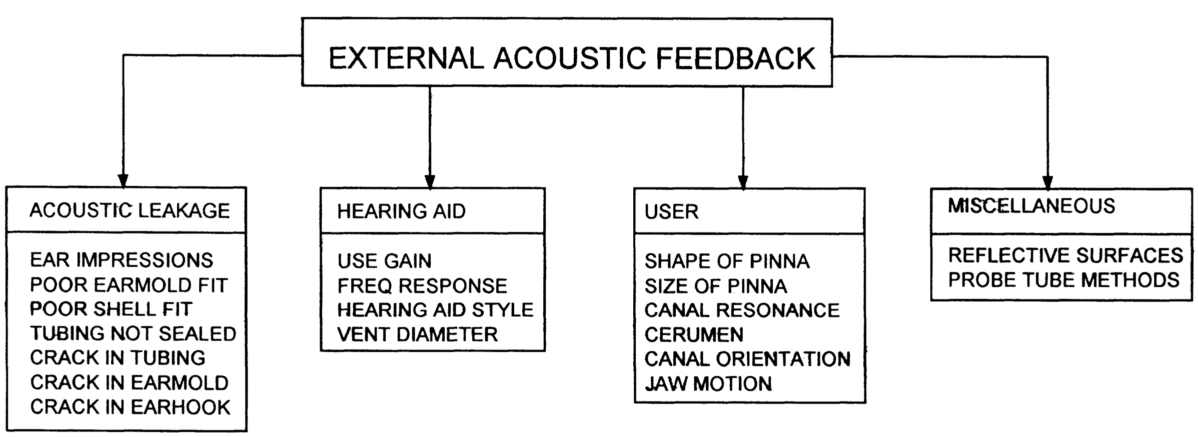

Figure 12. List of common causes of external acoustic feedback.

2. Inappropriate probe tube measurement technique.

A. Acoustic Leakage

1. *Inaccurate ear impressions*

The first two causes of external acoustic feedback have been separated into two headings even though they are related to the same general cause of feedback. The first factor relates to the need for a very accurate impression of the ear canal. The second factor is obviously closely related to the first factor. An inaccurate impression will likely lead to a poorly-fitting shell or earmold which, in turn, may lead to acoustic feedback due to the creation of a number of slit leaks.

One of the easiest methods to control acoustic feedback is to ensure a proper fit of the shell or earmold in the ear canal through an accurate impression. Poor impressions of the ear canal usually result either from poor impression technique or from subsequent dimensional distortion of the cured impression. Proper ear impression techniques are not within the scope of this text and may be learned elsewhere. An outline of the correct procedure for taking ear impressions is given in several publications, such as Morgan (1994) or Microsonic (1995). However, reading these sources does not substitute for hands-on practice in workshops and dispensing offices.

Hearing aid manufacturers and earmold laboratories usually make very accurate reproductions of the ear impressions they receive. However, they cannot compensate for a poorly-made impression or for the shrinkage and warping associated with some impression materials. Further insight into the problems that can occur with impression stability may be found in Agnew (1986a; 1986b). Newer materials, such as silicone, may be used to produce an accurate and long-lasting impression of the ear canal.

One suggested solution for particularly troublesome feedback problems is to use a three-stage impression technique developed by Fifield at National Acoustic Laboratories in Australia. The technique is outlined in Skinner (1988) and is described here because of its possible usefulness, especially for users with profound hearing losses.

1. Stage I: an impression is made and allowed to cure. It is then removed from the ear canal and trimmed.
2. Stage II: the impression is coated with additional impression material that is mixed to be more liquid than for Stage I. The impression is then reinserted into the ear canal after the ear canal and concha have been lubricated with oil. This second impression is then removed after curing.
3. Stage III: additional liquid material is placed in the ear canal and concha. The impression from Stage II is then reinserted into the ear canal. This final impression is removed after curing and sent to the manufacturer.

2. *Poor fitting of the shell or earmold*

The second problem is that, even though great care may have been taken in making the impression, the shell or earmold does not fit properly in the ear. This results in slit leak which causes acoustic feedback. Before suspecting this as the cause, it is important to make sure that the wearer has seated the hearing aid or earmold properly in the ear. This may be a simple, but sometimes overlooked, cause of the problem.

Fit problems may also be inherent. For example, an ITE hearing aid may have been re-cased at the factory and does not fit. This would indicate another return for re-casing. Alternately, problems may gradually become increasingly worse. For example, some ear canals may enlarge due to long-term use of a hearing aid or due to normal growth of the ear canal of a child. A loose-fitting earmold may be temporarily sealed to the ear by coating the pinna and canal with a thin layer of petroleum jelly. Alternately, putty may be used to seal around the rim of the hearing aid where it meets the concha. If this temporary seal cures the problem, a remake is indicated. For very high gain hearing aids, additional build-up of the impression at the manufacturer may be requested in order to produce an appropriate fit. A summary of ways that one earmold laboratory has found to stop feedback related to the earmold is given in Castleton (1983).

For clinicians wishing to modify the shell or earmold in the office, coating and build-up materials are available from various manufacturers and earmold labs. Build-up material may either be soft or hard and is available from various sources. For cases requiring only a thin build-up, some clinicians use clear, non-allergenic fingernail polish, which is commercially available in several brand names and found in many drug and sundry stores. Another simple technique that has been described for canal build-up is the use of a canal sleeve. This is a soft plastic sleeve that is placed over the canal, then trimmed to fit and glued in place (Orton, 1980b).

3. *Tubing not properly sealed to the earmold*

If the connecting tubing between the earhook and earmold is not tightly sealed, acoustic leakage may occur with resulting feedback. One way to test for this condition is to place a finger over the tip of the earmold to tightly seal the sound bore with the gain control at the full-on position. If the feedback stops, then the problem is probably the fit in the ear, or some cause other than poorly-sealed tubing. If the feedback continues, remove the earhook and earmold, and place the finger tightly on the receiver nozzle. If the feedback stops, then the earhook, tubing and earmold should be investigated further.

If a poor seal of tubing to the earmold is suspected, it is easy to produce a temporary solution by adding putty or petroleum jelly around the tube or snap-in nubbin where it enters the earmold. If the oscillation ceases, the putty may be removed and the tubing sealed firmly to the mold with adhesive or the snap-in adapter replaced. Obviously this problem cannot occur with in-the-ear hearing aids because the earmold and acoustic tubing are absent.

4. *A crack in the earhook, tubing or earmold*

A crack in the earhook, tubing or earmold may be observed by inspection of these components under magnification or may be found via the acoustic test outlined in the last section. It is easy to substitute another earhook or tubing to determine if this is the cause of the feedback. If the acoustic tubing is discolored or feels hard to the touch, it should be replaced as a matter of course. Polyvinyl-chloride (PVC) tubing used commonly for acoustic tubing is susceptible to deterioration from skin oils and will eventually turn hard and brittle as the plasticizers, that are present to make the tubing soft and flexible, are leached out. It is unlikely that the problem will be a crack in the earmold; however, this can occasionally happen.

When dealing with power BTE aids, it is common to use *thick-walled or double-walled* tubing to prevent sound leakage through the tubing and reduce the probability of feedback (Valente, et al 1996). Such leakage may occur even if the tubing is undamaged. Flack et al (1995) showed a 2 dB increase in sound attenuation when #13 tubing was changed from standard (2.9 mm) to thick-walled (3.3 mm) tubing. Nolan (1983) reported a reduction of sound radiation between 8 dB and 10 dB at the higher frequencies. Experiments in our laboratory have shown that leakage through the wall of the tubing can be reduced by up to 5 dB, depending on the thickness of the tube wall. While these differences may appear as small, they could contribute significantly to feedback reduction for difficult fittings.

B. Hearing Aid Characteristics

1. *Excessive high frequency gain*

Acoustic feedback may be caused when fitting a hearing aid that has excessive gain at a potentially unstable frequency, which may violate the Nyquist Stability Criterion. This means that even though everything was done to create an appropriate fitting, there is still excessive gain at the unstable frequency. Potential solutions for this type of feedback will be found through reducing the overall or high frequency gain. This will be described later in more detail when electronic and acoustic solutions to reduce feedback are presented.

2. *Proximity of components due to hearing aid case style*

As hearing aids have become smaller, the potential for acoustic feedback has increased because the components are closer together and the acoustic pathway between the microphone and the receiver has become shorter. For example, BTE hearing aids provide better stability because they usually have a longer acoustic pathway between the sound outlet port in the ear canal and the microphone port behind the ear. Also, BTE hearing aids often have internal mechanical baffles and cavities to further isolate the microphone and receiver from producing internally-transmitted feedback. The advent of in-the-ear hearing aids has forced the microphone and receiver closer together which, in turn, has created a higher potential for feedback. As the size progression has advanced from ITE to ITC to CIC, the proximity of the microphone and receiver has decreased, which has further increased the potential for acoustic feedback.

To provide some perspective on progress in combating the problems of feedback, it is interesting to consider how advances in hearing aids have changed the presence of feedback. In the mid-1960's, a transition occurred from a market dominated by body-worn hearing aids to BTE hearing aids. At that time, designers had a difficult time obtaining 50 dB of gain without feedback in a large BTE case. As manufacturers became more experienced and newer methods of transducer isolation, mounting and damping were developed, the upper end of the possible gain range has expanded to 75 dB without the presence of feedback. The introduction of electret hearing aid microphones in the early 1970's had a significant impact on this ability to increase gain because of their reduced sensitivity to vibration (Killion, 1975).

Orton (1986) pointed out that as the trend towards in-the-ear hearing aids was moving from ITE aids toward ITC aids in the 1980's, feedback became an increasing problem. He stated that feedback problems in ITC aids accounted for a remake rate of approximately two-and-a-half times greater than it was for ITE hearing aids. Again, this forced the creation of newer design techniques in amplifiers and construction which significantly decreased remakes due to feedback.

As a decrease in hearing aid size further advanced to CIC hearing aids, the problems of acoustic feedback have again re-occurred in the 1990's. In a survey of 122 fittings performed by various clinicians, Voll and Lyons (1995) reported that 62% of CIC hearing instrument fittings required shell modification to eliminate feedback. In most cases, the feedback was related to slit leakage. In 83% of these cases, the addition of soft build-up material to the shell of the hearing aid solved the problem. In approximately 8% of the cases, hard build-up material had to be added to enlarge the hearing aid case. In the remaining 9% of the cases, an acoustic damper was added to the receiver tube to smooth a resonant peak in the frequency response.

3. *Large vent diameter*

Amplification of the low frequency energy of a wearer's voice in a completely or partially occluded ear canal can produce an unnatural sound called the *occlusion effect*. This is often described by the wearer as "echoing" or "hollow". Venting is usually required in sloping mild to moderate sensorineural hearing losses to reduce the occlusion effect and reduce low frequency gain where hearing may be in the normal range. However, the presence of venting can cause a feedback pathway out of the ear and thus result in acoustic feedback. This is a serious problem when using a wide vent or an IROS earmold. More complete descriptions of the occlusion effect have been presented elsewhere. Solutions to prevent or minimize the problem have been described by Killion et al (1988) and Revit (1992).

For hearing aid fittings requiring mild to moderate amounts of peak gain (up to about 40 dB), acoustic feedback is primarily determined by the presence of venting. For fittings requiring greater amounts of gain (40 dB to 70 dB, or higher), acoustic feedback is typically caused by a poor-fitting earmold producing slit leakage, because these fittings generally use a very small *pressure vent* (also called a *static relief vent* or *static pressure relief vent*), or no vent at all.

As an example, tests on a vent 3 mm in diameter and 12 mm long by Johansen indicate that 35 dB to 40 dB of usable high-frequency-average (HFA) gain should be possible with a BTE fitting (Lybarger, 1982). Similar estimates by Johansen indicate that the maximum gain for an open mold fitting should be about 30 dB to 40 dB. Lybarger (1982) has reported results that are in agreement with those reported by Johansen. An investigation into the effects of a range of vents typically used with custom full-shell ITE hearing aids has been presented by Tecca (1991).

One of the easiest ways to determine if changing the vent diameter will eliminate feedback is to use either the PVV (Positive Venting Valve) or SAV (Select-A-Vent) venting systems. Both of these systems use changeable inserts with various vent sizes that can be inserted into a pre-drilled vent channel in the shell or earmold. The SAV system is essentially the same as the PVV system, but has slightly softer insertion rings. The effects of different SAV diameters on the maximum usable gain of a power hearing aid has been reported by Kuk (1994).

If variable venting has not been installed in the shell or earmold, a vent may be temporarily reduced or closed using putty to determine whether reducing the vent diameter will eliminate feedback. If reduction of the diameter of the vent is successful, then the vent may be permanently modified using hard acrylic (methyl-methacrylate polymer) material that is available in small quantities as patch kits from various sources.

C. User Characteristics

1. *Sound reflection due to the shape and size of the pinna*

For some wearers of in-the-ear hearing aids, the shape and size of the pinna may provide a reflecting or resonant surface that may initiate or exacerbate acoustic feedback. Individuals with large curved pinnae that bend forward over ITE-style hearing aids may create reflections from slit leakage that are reflected back to the microphone. This is similar to cupping a hand over the ear. When this occurs, it is difficult to troubleshoot and cure. If pinna reflections are thought to be a problem, manipulating the pinna with the hearing aid in place while feedback is occurring may give some clues.

Since the various components that make up the whole feedback pathway are so closely interrelated, it is sometimes difficult to ascribe the problem to one particular cause. The true root cause may never be known. Also, since it is not possible to modify the patient's anatomy, the only real cure for this situation may be a modification of the hearing aid shell or earmold by the factory. Examples of this type of modification are to angle the faceplate differently; to provide a tighter fit; to lengthen the canal section of the hearing aid; or to alter the vent.

2. *Excessively high ear canal resonance*

The average unaided human ear canal resonance for a sound with frontal incidence is approximately 17 dB (Shaw, 1980). Occasionally a clinician may encounter an individual who has a canal resonance which may be significantly higher or lower than 17 dB. For example if the canal resonance is 25 dB to 30 dB, this can interact with the high frequencies, acoustically amplifying them and causing feedback. The evidence for this occurrence appears to be anecdotal, and it can be argued that unaided canal resonance does not relate to occluded canal resonance in the aided condition. However, if this condition is suspected, any problem may easily be observed by using probe tube measurements. If excessive resonant peaks in the frequency response are observed, then appropriate modification of the high frequency amplification should be made to reduce high frequency gain.

3. *The presence of excess cerumen in the ear canal*

One major cause of feedback is the presence of excess cerumen in the ear canal. A hard lump of wax in the ear canal may provide a surface that is acoustically reflective and exacerbate feedback through the vent. The solution is to inspect the external auditory canal with a standard or video otoscope and remove the cerumen.

4. *Orientation of the canal tip towards the wall of the ear canal*

Feedback may also be caused by an orientation of the hearing aid sound bore in the ear canal such that sound is reflected back out through the vent from the sides of the canal. This may be a particular problem if the sound port is pointed directly at one of the bends in the ear canal. In this case it may be necessary to shorten or lengthen the canal portion of the hearing aid or earmold, or re-orient the direction in which the sound outlet points into the canal. It is important to ensure that the sound outlet points at the eardrum and not at the wall of the ear canal.

A separate, but related, problem may occur with ITE hearing aids using extended receiver tubing. In this case the extended tubing may be so long that it contacts the canal wall, bends and collapses, thus resulting in feedback. The solution may be to shorten the receiver tubing to prevent the problem from occurring.

5. *Mandibular motion causing slit leak*

Movement of the jaw can often initiate feedback. This is especially true in cases of sub-oscillatory feedback. Motion of the mandibular condyle in the mandibular fossa of the temporal bone, due to chewing, speaking, yawning or grimacing, can produce enough jaw movement to move an earmold or shell in the ear canal and cause acoustic feedback as a result of the creation of slit-leak due to this motion. This type of feedback may appear intermittently as the jaw opens and closes. Contrary to most feedback modifications which require lengthening the canal portion of the hearing aid or earmold to provide a better seal in the ear to prevent feedback, shortening the canal section of the hearing aid in some cases of mandibular joint problems may prevent jaw motion from moving the hearing aid or earmold and, in this way, possibly prevent feedback (Mullin and Barr, 1980).

This problem may also be reduced by a variant of the impression-taking technique. Morgan (1994) has recommended having the wearer talk, turn the head, smile and chew while impression material is curing in the ear. This allows the impression to reflect the dynamic conditions of the ear canal and reduces the probability of acoustic feedback during subsequent jaw movement.

D. Miscellaneous

1. *The presence of nearby reflective surfaces*

Reflective surfaces near the ear may initiate intermittent acoustic feedback. Examples of this situation for the wearer may be putting on a hat, having a full hairstyle or being close to a wall or other hard reflective surface. Since this type of acoustic feedback is situational, it may be fairly easy to diagnose the problem by asking the wearer to describe the situations in which feedback occurs.

2. *Inappropriate probe tube measurement technique*

As a final note, it is important to mention one problem that may be encountered when making probe tube measurements. Making such measurements improperly may initiate acoustic feedback during the process of fitting a hearing aid, where feedback would not normally be present while the hearing aid is being worn in everyday listening situations.

The practice of placing a probe tube between the earmold or the case of a hearing aid and the wall of the ear canal can cause acoustic feedback due to the inadvertent creation of slit leak. The other undesirable result of placing the probe tube in this manner is compression of the probe tube due to being tightly pressed between the hearing aid and the canal wall. This can result in invalid probe tube measurements. Instead, it is preferable to use a probe-tube vent. This is an additional small-diameter vent that is added to the earmold or shell during manufacture to accommodate the probe tube. This vent is plugged after the fitting is complete. When ordering a probe-tube vent from the manufacturer, it is important to specify the type or size of the probe tube that will be used since the vent is made only large enough to accommodate the outside diameter of the probe tube, in order to prevent additional acoustic leakage.

CAUSES OF INTERNAL ACOUSTIC FEEDBACK

The second general category of acoustic feedback encompasses problems occurring due to malfunction of the hearing aid. These malfunctions are generally technical problems which require the aid be returned to the factory for service. However, a clinician can sometimes perform minor repairs in the office to resolve a problem without having to return the aid to the manufacturer.

This issue does not substitute for repair workshops or modification seminars. However, the general principles of repairs will be discussed. The specific techniques for modifications and repairs will not be presented in detail here since these are rather specialized techniques that are best learned while attending modification workshops. For those who wish to learn by trial-and-error, further practical information and techniques for hearing aid modifications and repairs may be found in Orton (1980a; 1980b; 1981), Agnew (1985), and Riess and Guthier (1986).

The first level of analysis is to determine whether the problem is due to internal or external feedback. One quick method to determine if the cause of the feedback is internal is to turn the gain control full-on and to place a finger over the microphone opening tightly enough to acoustically seal it shut. If the feedback stops, then the cause is not due to a problem within the hearing aid and external causes must be investigated. If the feedback persists, the problem is probably internal and the hearing aid may require a return to the factory for repairs. This diagnostic process is outlined in Figure 13. It may be helpful to use a listening tube or stethoscope to hear the feedback clearly while performing this test.

Internal acoustic feedback tends to be caused by mechanical problems within the hearing aid. This includes problems such as mechanical feedback through the case walls, a dislodged receiver or microphone, a disconnected receiver sound tube producing acoustic leakage inside the hearing aid, or a receiver tube with a tear. A more subtle cause may be a pinhole leak in the receiver sound tubing leading from the receiver to the outside of the case. Such a leak can occur due to vigorous cleaning of the receiver tube with a wax removal tool.

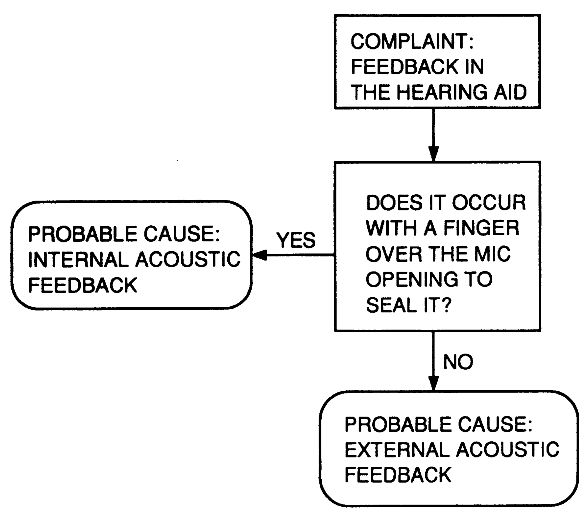

Figure 13. Flow chart for differential diagnosis of internal versus external acoustic feedback.

Mechanical feedback

Ridenhour (1988a; 1988b) theorized that mechanical feedback originating from the receiver may be transmitted back to the microphone through the plastic case that houses the hearing aid. Ward (1989) described a method of modifying the plastic case material to damp these potential vibrations by the addition of glass microspheres. Evidence that this is a real problem is inconclusive. Studies by Williams and Gutnick (1990) and Letowski et al (1992), comparing traditional and modified material, found no objective evidence to support any differences in the materials used to case hearing aids as far as promoting or eliminating mechanical feedback.

To evaluate the effects of mechanical vibration on feedback, engineers in our laboratory measured the transmission of vibrations through various ITE hearing aid cases. Induced case vibrations were measured using laser-based holographic measurement techniques. The results showed that mechanical vibrations induced in the case were less than 0.005 micron (5×10^{-9} meters) in amplitude with input sound levels of up to 100 dB SPL and during saturated amplification caused by acoustic feedback. Further measurements revealed that inherent mechanical resonances induced in the hearing aid case did not occur at the same frequency as did those for frequencies which caused acoustic feedback. It was concluded that the potential for mechanical feedback via this mechanism to cause significant problems was small when compared to other forms of feedback.

What is often thought to be mechanical feedback is actually feedback due to acoustic leakage directly through the sides of the receiver case or through the sound port tubing between the receiver and the hearing aid case. Vibrations induced in the sides of the receiver housing make the receiver case act as a miniature loudspeaker and produce an audible sound. The acoustic vibration and leakage inside the case can produce very high levels of sound inside the hearing aid case. This sound can then leak back out of the case to the microphone causing acoustic feedback.

There are two forms of mechanically-induced feedback that are commonly observed in hearing aids and their effects are difficult to differentiate since they have essentially the same cause. One occurs when the receiver comes physically into contact with the hearing aid case, the microphone housing or circuit components inside the hearing aid case. This in turn transmits mechanical vibrations to the microphone through direct mechanical linkage. An alternate acoustic embodiment of the problem may occur when the receiver is in contact with the case, but is not touching other internal components. The mechanism in this situation is that vibrations from the receiver induce vibrations in the case wall that result in an acoustic radiation that leaks back to the microphone and initiates feedback (IRPI, 1987). These two problems are not as common in BTE hearing aids as they are in ITE hearing aids. BTE hearing aids contain a more rigidly-defined internal mechanical structure and assembly methodology that usually includes cavities in the case to mechanically isolate the microphone and receiver. The problems occur more often inside a custom in-the-ear hearing aid due to the unavoidable proximity of components inside the shell. These problem may be avoided or corrected by ensuring that the receiver does not directly touch the hearing aid case or other internal components.

At one time acoustic feedback caused by mechanical vibration was more common than it is now due to the higher sensitivity to mechanical vibration of the older types of microphones. Piezoelectric ceramic microphones, which appeared in hearing aids in the late 1960's, were especially sensitive to vibratory feedback. However, this type of microphone is no longer designed into modern hearing aids. The occurrence of mechanical feedback has all but been eliminated by the use of modern electret microphones because these microphones have a sharply reduced sensitivity to mechanical vibration.

Hearing aid problems that may be associated with internal feedback are listed below and reported in Figure 14. While reading this section, the reader may also wish to refer to Figure 15 which is a diagram of the internal components of

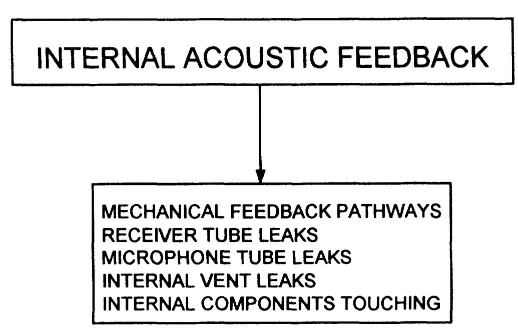

Figure 14. List of common causes of internal acoustic feedback.

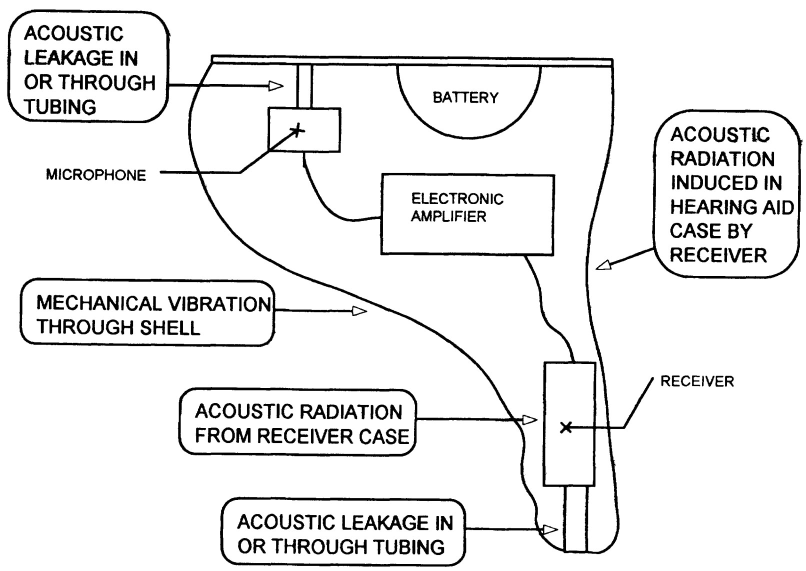

Figure 15. Cutaway side view of an ITE hearing aid showing potential internal feedback leakage pathways.

an ITE hearing aid showing possible sources of internal feedback.

1. Receiver tube leaks.
2. Microphone tube leaks.
3. Internal vent defects.
4. Internal components contacting one another.

1. *Receiver tube leaks*

Various mishaps can occur to the receiver sound tubing. The tube may become detached where it is glued to the receiver nozzle in a BTE hearing aid or to the case of an ITE hearing aid. This means that the receiver and tubing falls back into the hearing aid case. In this situation, sound will feed directly back to the microphone from the inside of the case. Alternately, the tube may become detached at the receiver end. Problems like this occur most frequently because the hearing aid was dropped. Even a drop from table-height onto a carpeted floor will produce very high impact forces on the receiver that can dislodge it from its proper mounting. Obviously, a fall onto a hard floor would be worse.

This condition may be observed and confirmed by inspection of the canal tip, or the point where the earhook connects to the BTE case, with a microscope or magnifying glass. If the end of the tubing is visible through the hole in the end of the canal tip, sometimes the problem may be corrected by grasping the end of the tube with narrow tweezers and carefully pulling the tube back through the case wall. Manipulation of the tubing has to be performed carefully in order to prevent tearing or puncturing the tubing with the tweezers, since any puncture or tear will immediately initiate feedback.

When the tubing is in place, a thin bead of cyanoacrylate adhesive (generically called super glue) or RTV (room-temperature-vulcanizing) silicone rubber may be used to hold the tube permanently in place. Cyanoacrylate adhesives cure within minutes, but some brands of glue may be unpredictable in performance, especially when gluing silicone rubber. RTV adhesives are more reliable for gluing silicone tubing. However, full adhesion usually takes several hours to cure to a tacky state and then an additional 12 to 24 hours is required to complete the full cure. Only the non-corrosive electronic grades of RTV silicone should be used in order to prevent corrosion of the transducers from acetic acid fumes produced by the curing of most RTV rubbers.

Receiver sound tubes are generally made of silicone rubber. The silicone rubbers tend to be either translucent, opaque white, opaque buff or opaque red. Silicone rubber tubing is highly resistant to skin oils and perspiration. In addition, silicone rubber tends to hold its shape very well. Occasionally butyl rubber or neoprene rubber sound tubes may be encountered. These types of tubing are generally a dull opaque black color. However, color should not be used as an exclusive identifier for either silicone or the other types of tubing. Butyl and neoprene rubber tubes are more common in older hearing aids. These types of rubber are extremely susceptible to deterioration by skin oils and it is common to find that these types have deteriorated to the point where there are tears, pinholes and thin spots in the tubing, or that the bore of the tubing has collapsed altogether. If minor repairs are made in the office, suspect black sound port tubing should always be replaced with new silicone tubing to prevent future feedback problems due to tubing deterioration.

Another source of feedback from receiver tubing is one that cannot be corrected easily by the clinician. This problem is due to sound radiation inside the hearing aid case that leaks through the walls of the receiver tubing. This problem typically only occurs with extremely high-gain hearing aids and is not necessarily a hearing aid malfunction, but may be due to a design weakness. In a well-designed hearing aid, this problem should not occur if the manufacturer uses suitable thick-walled tubing for the connecting sound tubing. Of course, failure of the tubing, such as thin spots or pinhole leaks can lead to the same symptoms. The solution in either case is to return the hearing aid to the manufacturer for repair.

Tears and holes in the receiver tubing commonly occur due to inappropriate techniques used for cleaning wax from the receiver tubing of in-the-ear hearing aids. Proper instruction is required for the wearer to accomplish adequate cleaning without damaging the receiver tubing.

2. *Microphone tube leaks*

A similar tubing problem may occur with microphones and microphone tubing. If the microphone uses a rubber sound tube, the tube may become detached at either the case or the microphone sound port end. If the microphone does not use a sound tube, but is placed in a pocket in an ITE faceplate, it can still become detached or dislodged and fall back into the electronics cavity of the hearing aid.

The cure for this problem is the same as was explained for the receiver tubing. If the detached microphone can be seen and maneuvered back into place in the BTE case or ITE shell, it should be performed as it was for the receiver tube. Usually this problem can only be corrected by opening the hearing aid case and physically re-attaching the tube.

3. *Internal vent defects*

Internal acoustic feedback may be caused in in-the-ear hearing aids by an internal defect, such as a pinhole leak or a crack in the wall of the vent channel. This allows sound to leak from the vent channel back into the cavity inside the hearing aid and thus to the microphone. This undesired acoustic leakage may be caused by a crack, a pinhole, or a thin spot in the material making up the wall between the vent channel and the electronics cavity of the hearing aid. This allows acoustic leakage from the receiver to exit the vent and travel back to the microphone to initiate acoustic feedback. This situation is different than cavity venting, which will be described later. The cavity vent is intentionally tuned to create a filtering effect on the frequency response. The type of defect described here results in undesired leakage that can induce acoustic feedback.

If this problem is suspected, the following temporary analysis can be tried. Plug the faceplate end of the vent with putty and place the hearing aid in the wearer's ear. If the feedback stops, the cause is either a crack in the vent channel or is due to acoustic feedback through the vent. If the feedback continues, it may be due to slit leakage which may be eliminated by sealing the hearing aid to the ear with petroleum jelly or putty. If the feedback still continues after this, it may be due to internal feedback through the vent channel.

Next, remove the hearing aid from the wearer's ear. Now remove the putty from the faceplate end of the vent, and place it on the canal end of the vent. Replace the aid in the wearer's ear and listen for feedback. If the feedback stops when the faceplate end of the vent is closed, but squeals when the canal end is plugged, the chances are good that there is a crack in the vent channel leading to the inside of the shell. If the feedback stops, cracks in the vent are not present.

Sometimes a crack, thin spot or hole may be seen by looking down the vent channel with a microscope and strong illumination. If such a crack or hole is observed, it may be possible to reach and repair it using shell-patch material available from various manufacturers.

4. *Internal components contacting one another*

Sometimes internal acoustic feedback may be caused by the receiver touching, or pressing against, the microphone, the case or some other internal component of the hearing aid. This can cause feedback via mechanical coupling. This mechanical pathway is more common in in-the-ear hearing aids, since the components are assembled in one large open central cavity, than in BTE products. These latter hearing aids typically have separate cavities for the microphone and receiver to isolate them. Mechanical feedback may also be transmitted through stiff wires used to connect the transducers to the electronic circuit.

If this problem is suspected, it may be possible to alter the position of the receiver and prevent feedback without having to open up the hearing aid case. This is done by carefully grasping the end of the receiver sound tubing with tweezers and either pulling the tubing slightly out of the case, or by rocking it gently from side to side. Inserting a wooden toothpick into the tubing where it exits the sound coupler in a BTE or where it connects to the canal tip for a custom in-the-ear product may serve the same purpose. Sometimes the receiver may be moved away from whatever mechanical pressure point is causing the problem and the feedback will stop. If the battery compartment is the unenclosed type, it may be possible to reach through the opening with a toothpick or pair of tweezers and push gently on the microphone or receiver. In this way, it may be possible to reposition one or both of the transducers to prevent feedback. If these simple steps do not solve the problem then the hearing aid should be sent out for service.

Manipulation of the tube may or may not solve the problem. In fact, even if this does solve the problem, there is no guarantee that the problem will be solved permanently. These techniques may be used in the case of an emergency repair; however, a more reliable solution is to return the hearing aid to the factory for service.

As with the other techniques mentioned above, great care should be taken not to puncture or tear the sound tube. This is generally a soft silicone rubber material that can be easily damaged by sharp tools.

As can be seen, correction of some of these problems goes beyond the category of simple repairs and progresses into advanced repairs. Clinicians who have had training or experience in opening and resealing hearing aid cases may wish to tackle these repairs themselves. However, for most clinicians, these problems should signal a factory repair. Remember that opening the case in the office may void the manufacturer's warranty.

CAUSES OF OTHER UNDESIRED AUDIBLE SOUNDS IN HEARING AIDS

The preceding sections have described the causes of acoustic feedback from the point of view of the hearing aid and the fitting of the hearing aid. As stated in the introduction, it is important to also discuss other types of undesired audible sounds that occur in hearing aids. This is important in order to distinguish various sounds from each other and to assist the clinician in separating symptoms and causes when troubleshooting oscillation and feedback problems. All of these types of problems, whether caused by feedback or not, may manifest themselves as various types of audible noises, such as "howling", "whistling", "sizzling", "roaring" or "buzzing sounds" that come from the hearing aid. Often these sounds are described generically by the wearer as *feedback*.

Unfortunately, from the clinician's viewpoint, there is usually little that can be done to the hearing aid to correct problems within this category. In almost all cases it is either necessary to remain clear of an environment that causes these problems, or to return the hearing aid to the factory for service. Specific recommendations are made in each of the following sections.

Other forms of oscillation in hearing aids that may be encountered are listed below and in Figure 16. These include:

1. Electrical feedback
2. Electromagnetic feedback
3. Electromagnetic pickup
4. Class D output stage oscillations
5. Oscillations due to battery problems

1. *Electrical feedback*

Up to this point the descriptions of feedback and oscillation have concentrated on acoustic feedback because this type of feedback causes the most problems for clinicians. However, in order to present a complete picture, it is necessary to include a description of electrical feedback since this type of feedback may manifest itself in a fashion which is very similar to acoustic feedback.

Electrical feedback is caused by a small portion of an amplified electrical signal leaking back into an earlier stage of the amplifier, where it is continuously re-amplified until circuit instability oc-

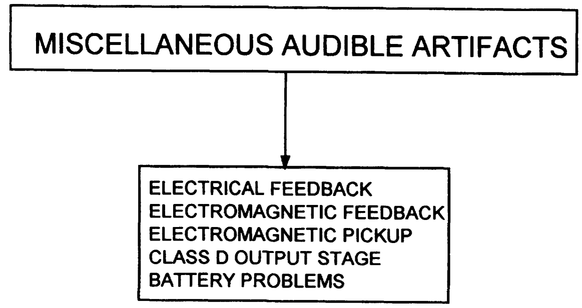

Figure 16. List of common causes of miscellaneous audible artifacts.

curs. This type of feedback results from a common coupling through some undesired electrical impedance within the circuit. Commonly this is caused by a high resistance ground point in the circuit, or by common coupling through the small (but finite) impedance of the battery.

Electrical feedback may lead to a variety of noises heard by the wearer. These noises range from a high-pitched "whine" or "roaring" to low frequency "puttering sounds", (commonly referred to as *motorboating*). Electrical feedback sometimes manifests itself as a very high-pitched tonal sound. This usually has a different audible characteristic than acoustic feedback, in that it is often heard as a constant high-pitched whine rather than a shrill squeal. Unfortunately, electrical feedback cannot be corrected in the field and requires a return to the factory for service.

2. *Electromagnetic feedback*

Electromagnetic feedback is very similar to acoustic feedback; however, in the case of electromagnetic feedback, the radiation and feedback pathway is electromagnetic instead of acoustic. Sometimes it has been termed *magnetic feedback*. However, the term *electromagnetic feedback* is a more precise name since the phenomenon is caused by radiated electromagnetic energy rather than by magnets.

Electromagnetic feedback is primarily caused by stray electromagnetic radiation transmitted from the receiver. This occurs because the electrical current passing through the coil inside the receiver causes low-level electromagnetic radiation similar to a miniature radio transmitting station. This energy leaks back into another part of the circuit causing squealing to occur.

When electromagnetic feedback occurs, the point of reception of the feedback is often the telecoil. This coil is designed to pick up electromagnetic energy radiated from the telephone handset. If the coil also picks up stray electromagnetic radiation from the hearing aid receiver, feedback may occur. In this way, particularly if the hearing aid gain or telecoil sensitivity is high, interaction between the telecoil and receiver occurs causing feedback. If the symptoms of feedback are eliminated when the gain control setting is reduced or when the telecoil is switched off, then the cause of the feedback may be electromagnetic. This problem is the result of either a poorly-designed hearing aid or a malfunction of the circuit. The only solution is to return the hearing aid to the manufacturer for service.

3. *Electromagnetic pickup*

A form of electromagnetic pickup that may be confused by the wearer with feedback and is often described as *feedback*, is *electromagnetic pickup* by the hearing aid from various sources of radiation. This is frequently the situation where the wearer complains of occasional "feedback" when in specific environments. Often this is not acoustic feedback, but is some form of electromagnetic pickup that may resemble feedback. Common environments where this may occur include computer monitors or video terminals, certain automobiles when the engine is running, electrical machinery, airports, automatic door openers, cellular telephones and motion detectors.

This problem usually manifests itself as an irritating high frequency whine that is present when the wearer is in a particular environment. When the wearer is away from the particular environment, this noise disappears. One way to confirm this problem is for the clinician to ask the wearer to keep a log of situations where and when the noise occurs, along with a detailed description of the noise.

For example, the following real-life cases were found to produce various noises:

1. The wearer observed a varying squealing noise when approaching a computer monitor in the office. This is an example of electromagnetic pickup of the horizontal sweep oscillator in the monitor.
2. The wearer observed a high-pitched tonal whine when riding in a particular automobile. The pitch of the whine changed as the car accelerated and decelerated. This is an example of electromagnetic pickup of a particular car ignition, which increased in pitch as the car accelerated.
3. The wearer observed a humming noise when riding in a certain elevator. This was caused by inductive pickup of the motor powering the elevator.
4. The wearer, who lived near an airport, observed a periodic high-pitched squeal. This was caused by the ground radar at the airport periodically sweeping around and being detected by the hearing aid.
5. The wearer observed a squealing when visiting an art museum. This is an example of the hearing aid picking up security motion sensors.

Although these problems initially appeared to be acoustic feedback problems, in reality they were examples of the undesired detection of energy that could not be simply resolved. Many other situations can be envisioned.

Though all these wearers complained of feedback, the causes of the problems were the result of environmental situations which were beyond the wearer's control. Should there be some offending environments, such as the ones described above, the only solution may be for the wearer to either stay out of those environments or turn off the hearing aids while in the environment.

One particularly troublesome source of radiation for hearing aid wearers is the noise generated by cordless telephones and by digital mobile cellular telephones (Joyner et al, 1993). European cordless telephones use a 2 millisecond switching rate with a power of 100 mW. The radiation from this switching rate creates a 500 Hz square wave (frequency = 1/time = 1/(2 millisec) = 1000/2 = 500 Hz) that is picked up and detected by the hearing aid. The GSM (French: *Groupe Speciale Mobile*; English: *Global System for Mobile*) digital cellular telephone system which is in widespread use in Europe, Canada and certain Asian countries produces an interference frequency of 217 Hz, and has a radiated power of 1 or 2 watts, depending on the country of use.

Much of the radiation from these systems is picked up by the wires of the microphone acting like small receiving antennas. Thus this interference goes beyond telecoil use of the hearing aid. Interference may be picked up while using the microphone, and may be picked up from as far away as several feet. This radiation is then detected and amplified by the hearing aid circuitry. The square waves produced by the switching produce many harmonics across the frequency range and produce an irritating buzzing noise in the output of the hearing aid (Hahn et al, 1993). This has been effectively demonstrated on a videotape comparing GSM, TDMA (time division multiple access) and CDMA (code division multiple access) telephones by Killion (1996). As of this writing, no satisfactory solution has been found and hearing aid wearers who wish to use GSM cellular telephones may be unable to do so. TDMA telephones, which may become the primary direction of technology in the US, have a lower radiated power level and correspondingly lower levels of interference. CDMA telephones use spread spectrum techniques and may also provide reduced interference problems.

4. *Class D output stage oscillations*

A less common form of undesired sound generation that is sometimes mistakenly called *intermittent feedback* is the presence of beat frequencies from a Class D receiver. This problem only occurs presently in Digitally Controlled Analog (DCA) hearing aids that contain two oscillators (clocks), one for the internal circuit and one in the integrated Class D receiver. This problem could also occur with future Digital Signal Processing (DSP) hearing aids. Those unfamiliar with the principles of DCA and DSP hearing aids may wish to consult Staab (1985), Conger (1990), Sammeth (1990), Staab (1990), Agnew (1991), and Sandlin (1994), for additional information.

This particular problem occurs if clock leakage from the DCA circuit bleeds into the electrical signal path at the output of the hearing aid. At the output amplifier, the DCA circuit clock leakage combines with the integrated receiver output stage clock. Since the two clocks are not synchronized, they beat together and produce audible difference tones. The typical acoustic symptom of this problem is a low-level sound that changes pitch while the wearer listens. At times the sound will seem to disappear as the clock frequencies or their harmonics drift and the beat frequency approaches zero. Then a low-pitched sound will appear and slowly rise, then reverse and become inaudible again. This cycle keeps repeating itself. The wearer typically reports intermittent feedback. This sound has been nicknamed the *Star Wars phenomenon*, as the varying pitch of the oscillation sounds like a sound effect from a science fiction motion picture. If symptoms like this occur in a hearing aid, the only solution is a return to the factory for repair or modification.

This phenomenon was occasionally reported when the first integrated Class D receivers were used in programmable hearing aids. It was generally the result of improperly designed hearing aids and is seldom now reported. The occurrence of this problem will probably decrease as more manufacturers incorporate Class D circuits directly into their integrated circuits and use one master clock for the entire hearing aid. An alternate solution is when integrated Class D receivers become available that synchronize their internal free-running clock with the rest of the hearing aid circuitry.

5. *Oscillation due to battery problems*

One unusual audible symptom that can fall into the category of hearing aid oscillation and malfunction is a problem with batteries. There are instances where a hearing aid will not function properly with a particular brand or batch of batteries. This may be caused by a marginally-adequate hearing aid circuit design, or by a particular batch of batteries that have characteristics not compatible with the original hearing aid design.

Due to their electrical and mechanical design, all hearing aid batteries have an internal impedance that increases with decreasing frequency. Hearing aid amplifiers are designed to remain stable with anticipated maximum values of this impedance. What may occasionally happen is that a particular brand or batch of batteries may have a higher impedance at a very low frequency than the hearing aid circuit was designed to tolerate. These troublesome frequencies usually lie below 1 Hz, often in the range of 0.1 to 0.3 Hz. This undesired increased impedance interacts with the hearing aid amplifier circuit, causing it to become unstable and produce audible symptoms.

Typically the symptom of battery problems is a constant very-low-frequency oscillation that sounds like a bubbling noise or like an outboard motor on a small boat, hence the common name of *motorboating*. The solutions in the dispenser's office are few. The obvious solution is to try a different brand or batch of batteries. If this solves the problem, it may be necessary to continue to use these batteries.

Sometimes this problem may be caused by a film of dirt or corrosion on a battery contact, which raises the electrical impedance at the contacts. This prevents the contacts from making a good electrical connection to the battery. Cleaning the battery contacts with a pencil eraser or a cotton-tipped applicator dipped in isopropyl (rubbing) alcohol may correct the problem. Alternately, bending the contacts slightly so that they exert more pressure on the battery may also resolve the problem. Bending the contacts must be done with care to prevent overbending or damage.

If these procedures do not resolve the problem then there is probably another reason for the malfunction, and the hearing aid should be returned to the manufacturer for service.

ELECTRONIC SOLUTIONS TO CONTROL ACOUSTIC FEEDBACK

Understanding the nature of acoustic feedback leads to an understanding of the numerous solutions to help solve the problem. One general method of resolving feedback is through electronic modifications via circuit adjustments. Electronic methods of feedback reduction include:

1. Overall gain reduction
2. Reduction of high frequency gain
3. Electronic damping of high frequency peaks
4. Bandpass filtering
5. Notch filtering
6. Frequency shifting
7. Phase shifting
8. Frequency warbling
9. Adaptive cancellation filters

1. *Overall gain reduction*

Since acoustic feedback is caused by a combination of phase angle and excessive gain at a critical frequency, one solution is to reduce the overall gain until the feedback ceases. Unfortunately, while this may eliminate the acoustic feedback, the overall gain may be reduced to the point that the gain provided to the wearer is inadequate to allow speech to be audible and intelligible.

Another problem may also occur. The wearer typically reduces the setting of the gain control to a position just below the point at which acoustic oscillation occurs. Often this is performed by the wearer cupping the hand over the ear and reducing the setting of the gain control to just below the point of acoustic feedback. Thus it is likely that the wearer has set the gain control to the point where the hearing aid has sub-oscillatory feedback. When an intense sound is amplified, this setting may force the hearing aid to be driven in and out of oscillation at feedback frequencies. The ringing that is produced adds a tinny distorted sound to the wearer's perception of amplification. See Cox (1982) for a further discussion of this problem.

2. *Reduction of high frequency gain*

Since acoustic feedback is unlikely to occur at all frequencies, one solution is to reduce the gain at those frequencies that are most likely to cause feedback. Acoustic feedback is typically a high frequency problem. Or, stated another way, the gain in the high frequencies at a critical 0° phase crossing is so high that acoustic feedback occurs. Thus, since acoustic feedback is caused by an unfortunate combination of high frequency gain and phase, one solution is to reduce the high frequency gain. Modification of the high frequency gain may be performed by adjusting a high frequency potentiometer or, if this is not available, by requesting a reduction of the high frequency gain from the factory when ordering the hearing aid.

The effect on the frequency response of progressively reducing the high frequency gain is

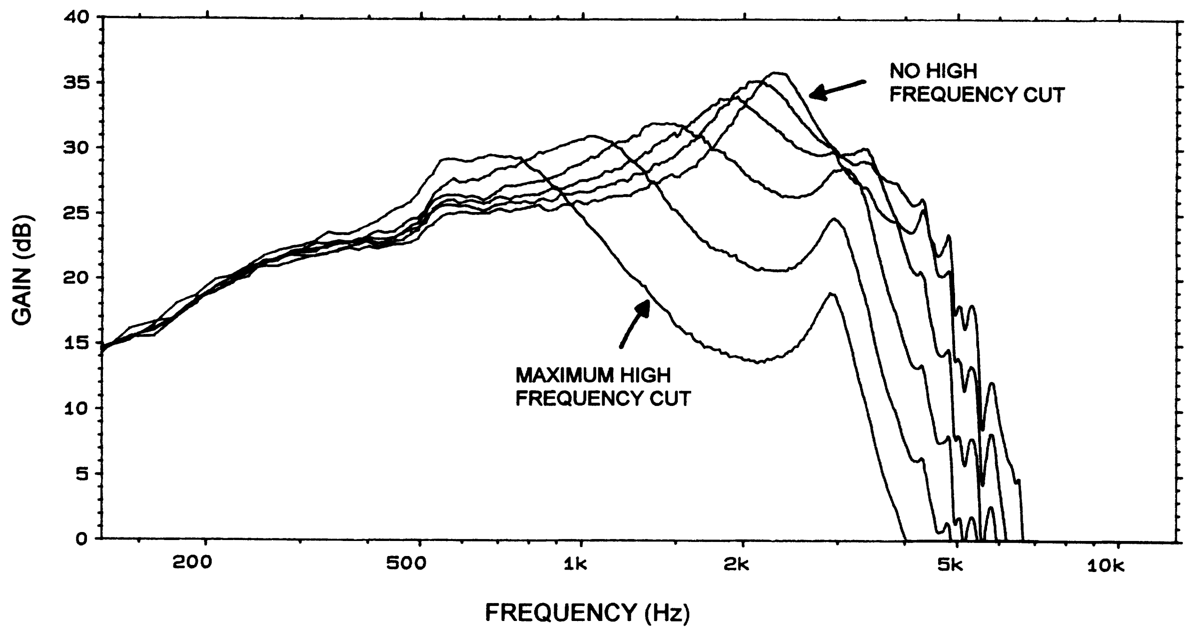

Figure 17. Six superimposed frequency response curves showing the progressive effects of a high frequency cut potentiometer on the response of a typical hearing aid.

shown in Figure 17. This figure shows six frequency response curves. As the high frequency gain is successively reduced, the characteristic primary receiver peak is progressively reduced in frequency and amplitude. The net effect is to lower the frequency of peak gain as well as reducing the overall high frequency gain.

Unfortunately, the reduction of high frequency gain leads to the same problem that occurs when overall gain is reduced. Elimination of acoustic feedback may require so much reduction of high frequency gain that the fitting may not provide adequate amplification in the high frequency region to meet the needs of the wearer. As a result, speech intelligibility may suffer. However, reduction of high frequency gain is often the compromise that has to be made to prevent feedback making the fitting viable.

A more critical problem may be that of *wasted gain*. An example of this is found with the wearer who has a profound hearing loss (80 dB HL or more) in the high frequency region and only a mild to moderate hearing loss below 1500 Hz to 2000 Hz. The clinician may attempt to fit a real ear target gain curve based on hearing threshold values to achieve the required gain. However, the problem is that the hearing loss is so great that the wearer cannot tolerate the prescribed amount of high frequency gain because it exceeds the loudness discomfort level. Additionally, this excessive amount of high frequency gain will lead to feedback. Thus, in order to avoid feedback and uncomfortable sounds, the wearer reduces the gain control setting. In turn, this reduces the amplification in the lower frequencies where it could be useful. Thus the additional gain is "wasted".

3. *Electronic damping of high frequency peaks*

An electronic feedback circuit developed to shift the primary peak of the receiver resonance to a lower frequency and to reduce its amplitude has been described by Preves et al (1986). The authors reported an increase in insertion gain of about 5 dB to 8 dB above 1000 Hz when using the circuit.

Powers and Sacca (1983) discussed the use of a *resonant peak control (RPC)*. This is a high-cut potentiometer that decreases the high frequency gain and shifts the resonant peak of the receiver to a lower frequency (see also Figure 17). Shifting the peak to a lower frequency decreases the gain at what could be a feedback frequency and thus reduces the potential for feedback. This is essentially another way of reducing high frequency gain to reduce feedback.

Teder (1992) has also discussed the effects of a high frequency roll-off circuit for reducing high frequency gain and damping the resonant peak.

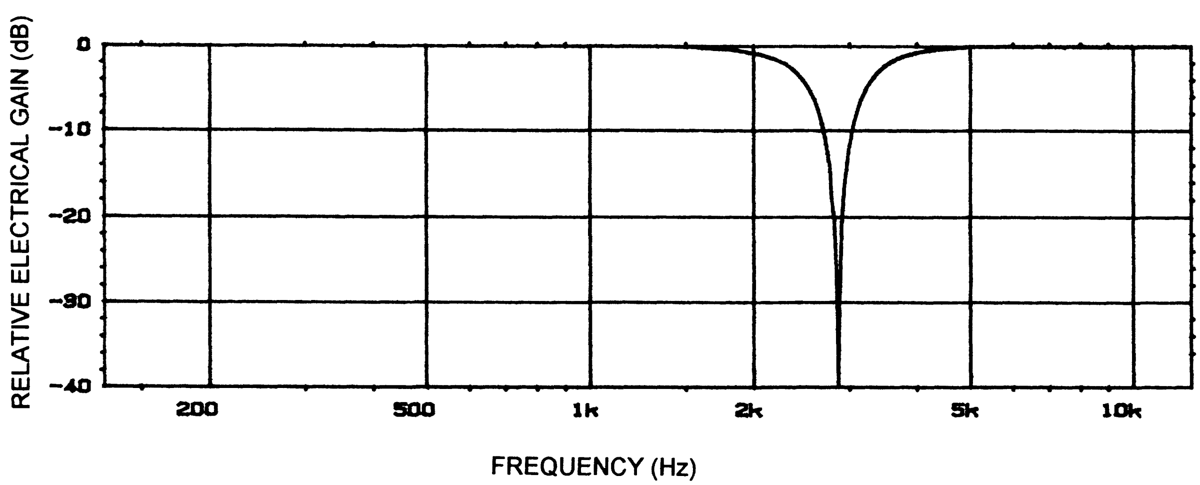

Figure 18. Electrical frequency response of a notch filter centered at 2800 Hz.

This circuit has been called a *Feedback Reduction Circuit (FRC)*.

4. *Bandpass filtering*

Patronis (1978) described a circuit consisting of 17 phase-locked loops and electronic attenuators to suppress feedback in public address systems from 50 Hz to 15000 Hz. The phase-locked loops detect feedback as a persistent single frequency that does not change in amplitude. The attenuators are gain controllers used to reduce the loop gain to below the Nyquist Stability Criterion of 1 at the potential feedback frequency or frequencies. Because of its complexity, a system requiring 17 controllers would be difficult to implement in a hearing aid and this has not so far been achieved.

Arcos et al (1995) have proposed a similar solution for a hearing aid where the circuit measures short-term and long-term energy in each of eleven bands and reduces the gain in any band that senses a steady-state high-energy signal (assumed to be oscillation from acoustic feedback) within that band. This has also not been implemented in a wearable hearing aid.

A similar, though less precise, approach to bandpass filtering may be taken when using a hearing aid with more than one channel of programmability. For example, a three-channel hearing aid is, in effect, a three-bandpass-filter hearing aid. Thus it is possible to adjust the parameters in one channel to reduce the gain at and around the feedback frequency, without substantially affecting the gains of the two other channels (Smriga, 1991). The drawback to this method is that a large area of the frequency response will be affected. Thus if more channels are available, the more precise the adjustment will be.

5. *Notch filtering*

Since acoustic feedback does not occur at all high frequencies, it is appropriate to reduce the gain at the specific feedback frequency and, at the same time, not affect nearby frequencies. This may be done with a *notch filter* that selectively reduces the gain at a particular frequency which may cause feedback.

A notch filter, more properly called a *stopband filter*, selectively removes a narrow band of frequencies around some pre-determined center frequency. The electrical frequency response of a typical notch filter, centered at 2800 Hz, is shown in Figure 18. As can be seen, the filter produces a dip, or notch, in the frequency response curve. The gain of the filter is unity (e.g. has a gain of 1, or 0 dB) at all frequencies except at the center frequency of the notch, where it provides 40 dB or more of attenuation. The shape of the transition region between the unity gain area and the notch frequency is determined by the hearing aid designer.

Likewise, the width of the notch, or the region of attenuation, can also be determined by the filter designer. The width of the notch filter, more correctly called the bandwidth of the filter, is described as the difference in the frequencies at which the response decreases by 3 dB from unity gain. In the case of the filter in Figure 18, the *bandwidth* is approximately 1000 Hz wide, which is the frequency difference between the two 3 dB-down frequencies.

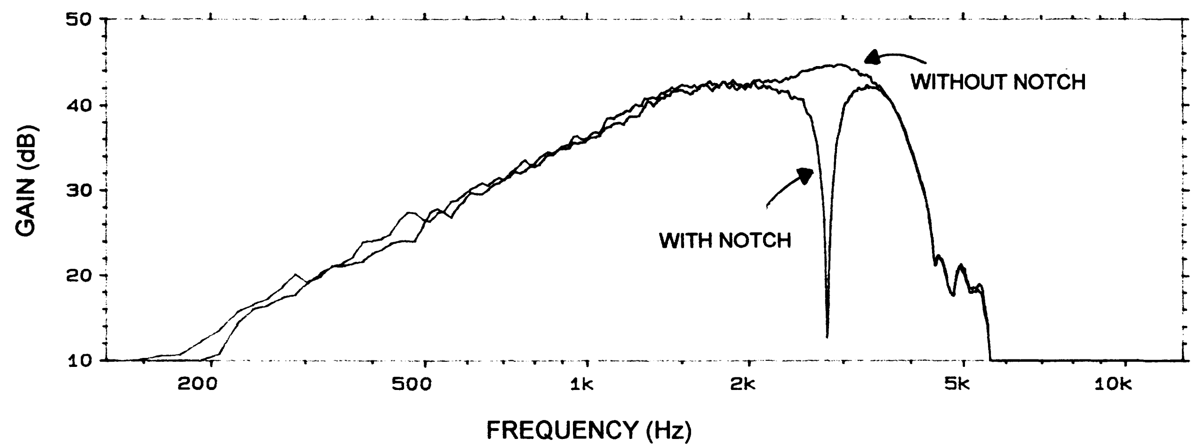

Figure 19. The effect on the frequency response of a hearing aid of adding the electrical notch filter of Figure 18 to the circuit.

Figure 19 shows the use of this notch filter in a hearing aid. This figure illustrates the original frequency response and the frequency response when the notch filter from Figure 18 is added to the circuit. It can be seen in Figure 19 that the original hearing aid frequency response maintains its shape, except around the center frequency of the notch.

Boner and Boner (1965) have described the use of notch filtering in public address systems to modify the gain at an acoustic feedback frequency. They reported between 7 dB and 18 dB of additional stable gain with the use of this technique.

A variable electronic notch filter for power hearing aids has been described by Agnew (1993). The center frequency of the notch filter can be adjusted from 1500 Hz to 5000 Hz via a potentiometer. In use, the trimmer is used to adjust the frequency of the notch to coincide with the frequency causing feedback. Real ear measurements have been used to observe the adjustment of the notch filter during hearing aid fitting on the wearer. In clinical trials, this notch filter provided 8 dB to 10 dB of additional gain in the lower frequencies. A smaller increase in gain was seen between 2000 Hz and 4000 Hz (Agnew, 1993).

Electronic notch filters have also been used in programmable hearing aids. One hearing aid has been described with a notch filter that has three different bandwidths (narrow, medium and wide) and a center frequency that may be programmed between 2500 Hz and 5500 Hz in 500 Hz increments (Agnew, 1992). During fitting, the notch is programmed to cover troublesome feedback frequencies. As well as being useful for feedback reduction, adjustable notch filtering may also be used to decrease resonant peaks in the frequency response to provide further flexibility for achieving prescribed real ear gain.

In real life, the use of a notch filter may suffer from two potential shortcomings. First, the use of a single notch may be inadequate to suppress all oscillations since feedback may occur at more than one frequency. If the notch is used to suppress the primary frequency, feedback may occur at a secondary frequency and again produce an audible problem.

Second, acoustic feedback is not a static phenomenon. A feedback frequency is determined by the acoustic conditions surrounding the hearing aid. In real life, users talk, chew, sneeze, cough, go near reflective walls, and generally move in and out of situations that change acoustic feedback pathways. These factors may alter the frequency causing the feedback. Thus, the use of a single notch filter becomes a compromise. If the notch is very narrow, it may not adequately suppress the feedback frequency over the expected range of frequency shift under different use conditions without subtracting frequencies important for speech recognition. If the filter is very wide, so that it can compensate for changing acoustic conditions without having to be continually reset, important information-bearing high frequencies may be eliminated, which leads to reduced intelligibility. Thus the use of a notch filter becomes a compromise between adequate feedback suppression and adequate intelligibility.

In an attempt to overcome this problem, Chen (1978) has disclosed a specific method for making an automatically-tunable notch filter for the suppression of acoustic feedback. Because of the complexity of the circuitry involved, this strategy is more suited to public address systems and has not yet been implemented in a hearing aid.

A further refinement of a single notch filter hearing aid is to incorporate several notch filters into the circuit. However, this rapidly approaches the limits of the amount of circuitry that can be practically incorporated into a hearing aid.

A system using two adaptive notch filters was implemented in a hearing aid in our laboratories. The circuit was designed to reduce feedback at any two separate frequencies. However, during use, it was noted that two notch suppressor circuits were inadequate to suppress all modes of feedback. When the primary and secondary feedback frequencies were suppressed, very often a third frequency would start to whistle. This meant that one of the suppressors would release from a suppressed frequency to suppress the third frequency. Of course, the first frequency would then again become audible. The circuit would switch back and forth between several feedback frequencies, being unable to suppress all frequencies at all times. The suppression circuitry was limited to two frequencies in order to maintain the circuit complexity at a level that could be reasonably implemented in a wearable hearing aid. Other factors that limited this circuit were the need to maintain the battery current drain at a reasonable level and to maintain the internally-generated circuit noise at an acceptable level. Because of these limitations, further development of this circuit concept was not pursued.

6. *Frequency shifting*

One of the strategies described for feedback reduction in public address systems is to shift the feedback frequency up or down by a small amount as it is processed through the amplifier. This shift means that the frequency of the signal being fed back and re-amplified is different each time that it passes through the amplifier. This procedure eliminates the feedback condition. Shifts of 5 Hz to 10 Hz have been proposed to achieve effective suppression.

Schroeder (1962; 1964) described the use of a frequency-shifting circuit inserted in the amplifier pathway of public address systems to cause the output frequency at a feedback frequency to be shifted to 5 Hz higher than the input frequency. This technique was used to stabilize the system at an unstable feedback frequency. Schroeder (1964) also commented that audible beating effects limited the practical available additional gain increase to about 6 dB.

Kryter (1975) has suggested that a frequency shift of 10 Hz will allow about 10 dB of additional stable amplification. It is not clear if the 10 dB of additional gain was based on theory or on actual measurement.

Bennett et al (1980) reported on an attempt to apply this principle to hearing aids. They reported that, while a fixed shift of 5 Hz was suitable for public address systems, acoustic feedback could still occur in a hearing aid when using this limited amount of constant shift. Thus, they increased the fixed shift to 30 Hz. This suppressed the feedback, but also introduced audible distortion because of alterations of the harmonic relationships in the original signal.

Their next attempt was to try a prototype hearing aid containing a progressive amount of frequency shift. At input levels likely to produce acoustic feedback, the amount of frequency shift was small. At higher input levels, a larger amount of shift was generated. Their data showed that under these conditions this hearing aid produced 6 dB to 15 dB of improvement in usable gain. The largest amount of increased gain occurred in the low frequencies. However, the authors also commented that at very high input levels a warbling noise could be heard in the output signal. If this strategy is pursued it will be important to test frequency-shifting suppression methods for any degrading effects on speech intelligibility (Egolf, 1982).

7. *Phase shifting*

Preves (1985) described laboratory experiments inserting a phase shifter into the forward signal path of a hearing aid. His experiments resulted in an increase in available gain of 15 dB over the uncompensated condition. When the gain of the circuit was adjusted to the edge of sub-oscillatory feedback, the transient response was improved and ringing was reduced when using this phase compensation. However, when the gain control was reduced to adjust the system below sub-oscillatory feedback, the transient response was the same with and without phase compensation. This concept does not appear to have been pursued to a commercial hearing aid.

Waterhouse (1965) has described a system to alter the phase of a circuit at a feedback frequency. Such a circuit, if adaptive, might work in a hearing aid, but has not yet been implemented.

Like many other potentially-promising anti-feedback strategies, the complexity of the circuitry involved has made it unattractive for practical hearing aid use.

8. *Frequency warbling*

Nishinomiya (1968) has suggested warbling a potential feedback frequency at a warble frequency of 5 Hz with a frequency deviation of 10 Hz. However, the author also comments on listener annoyance with the method and suggests that only the specific frequency where feedback is liable to occur should be warbled. This method has not been implemented in a commercial hearing aid.

9. *Adaptive cancellation filter*

This method of acoustic feedback suppression has been placed in a separate category because it has been successfully implemented in a wearable ear-level hearing aid.

Dyrlund and his colleagues (1994) have described the use of a *Digital Feedback Suppression (DFS)* hearing aid. Intended primarily for severe to profound hearing losses, this technology is available in a power BTE hearing aid using a single 675 battery. The circuit uses an adaptive digital Finite Impulse Response (FIR) filter to model and then cancel the external feedback path. In order to make the adaptive adjustment, the feedback path is measured using a continuous white noise test signal that is added to the signal pathway. The loop gain is analyzed by a method similar to that described in Figure 9 for the measurement of feedback loop gain. The white noise signal is used to determine the open loop gain of the system. Then a compensation signal is generated that has the opposite phase to the feedback signal. This is used to cancel the offending feedback signal. The processor has a sampling rate of 9.6 kHz and a resolution of 12 bits, or approximately 70 dB of dynamic range (Bisgaard, 1993). This technology has been described in detail elsewhere (Bisgaard and Dyrlund, 1991a, 1991b, 1991c; Bisgaard, 1993; Goodings et al, 1993; Smriga, 1993; Dyrlund et al, 1994).

Stated advantages of the DFS circuit are:

1. The circuit provides approximately 10 dB of additional usable real ear gain (Dyrlund et al, 1994; Smriga, 1993). A study of ten children, fitted binaurally, by Henningsen et al (1994) reported that 5 dB to 10 dB of additional usable gain in the low, mid and high frequency regions was available.

Stated difficulties with the system are:

1. The system does not store parameters in memory, but resets itself each time during power-up after the hearing aid has been turned off (Smriga, 1993), thus involving a time delay while the circuit stabilizes before use.
2. The audibility of the white noise test signal during measurement and activation of the feedback suppression cycle, which lasts about 2 to 3 seconds, may be objectionable to the wearer (Smriga, 1993).
3. Feedback caused by close reflections will be canceled. However, reflections greater than a half-meter away will not be canceled (Bisgaard, 1993).
4. Because the algorithm uses the criterion that input signals with long-term stationary characteristics are feedback, some warning alarms may fall into this category and may also be canceled (Engebretson, 1993).

Engebretson and French-St. George (1993) described a similar type of adaptive cancellation filter concept implemented in a prototype system. In their system, the characteristics of the acoustic feedback pathway were estimated using a known signal at the input of the hearing aid. An electrical signal with identical characteristics was generated and subtracted from the amplifier signal. This process was adaptive to compensate for any change in the feedback signal over time. A laboratory evaluation using a manikin showed a possible gain increase of 15 dB to 20 dB before feedback under ideal conditions. However, walking, talking, chewing and breathing reduced this 15 dB to 20 dB gain improvement. Trials of the system using nine subjects in a sound-treated room produced gain increases of 4 dB for the average listener (French-St. George et al, 1993). However, even this reduced amount of improvement can provide significant benefits to a hearing impaired wearer.

Similar concepts for adaptive feedback cancellation filtering have been described by Graupe et al (1988), Levitt et al (1988, 1989), Engebretson et al (1991), and Levitt (1993).

ACOUSTIC SOLUTIONS TO CONTROL ACOUSTIC FEEDBACK

Acoustic feedback may be reduced or eliminated by alteration of the physical conditions causing feedback. These might include modifications of the earmold or hearing aid case. For example, lengthening the canal portion of an earmold or in-the-ear hearing aid will change the feedback pathway and may be enough to reduce or prevent feedback. Reducing the size of the vent may reduce the amount of feedback at the critical frequency and may eliminate audible oscillations.

Useful information on types of earmolds, as well as explanations of the acoustic characteristics of earmold coupling and venting may be found in Libby et al (undated), Mynders (1980), Skinner (1988), Microsonic (1995) and Valente (1996).

Acoustic methods of feedback reduction or prevention include:

1. Elimination of the vent or reduction of its diameter
2. Ensuring a tight seal to the ear
3. High frequency gain reduction
4. Acoustic damping of receiver peaks
5. Use of dual receivers
6. Acoustic notch filtering
7. Modification of the canal tip
8. Physical separation of the microphone and receiver

1. *Elimination of the vent or reduction of its diameter*

Acoustic modifications to the vent to reduce the high frequency feedback path may be made by reducing the vent diameter via the use of variable vent (PVV or SAV) inserts; by reducing the vent diameter via the addition of acrylic material; or by placing a small piece of foam or lamb's wool in the vent.

The disadvantage of reducing the vent diameter is that wide venting may be required for a particular fitting in order to reduce low frequency gain for that fitting and to prevent the occlusion effect. In this case, a compromise must be made between the problems of feedback and other aspects of the hearing aid fitting.

In the case of earmolds, most vents should be parallel vents. Diagonal vents in earmolds are not recommended because diagonal vents have a greater tendency than parallel vents to cause feedback and reduce the high frequency gain (Microsonic, 1995). It should be recognized, however, that sometimes the only way that a vent can be included is to use a diagonal vent. This occurs when the diameter of the ear canal is very narrow, but venting is still necessary to avoid the occlusion effect and provide a reduction of low frequency gain.

2. *Ensuring a tight seal to the ear*

One of the simplest ways of reducing feedback is to ensure a tight seal to the ear in order to avoid slit leaks. Fit is particularly important with power hearing aids where venting is kept to a minimum in order to minimize potential acoustic feedback pathways. In these cases, slit leakage becomes the dominant feedback pathway.

Techniques to reduce slit leakage include building up the earmold to make a tighter fit. Additional strategies involve the use of a soft earmold or shell material that seals well to the ear. Soft material seals well, but is subject to more frequent deterioration than hard acrylic earmolds, since perspiration and oils in the ear leach out the plasticizers that make the material soft.

In mild cases of slit leakage, all that may be required to obtain a good seal in the ear is to lightly coat the pinna and outer portion of the ear canal with thin oil, such as mineral oil or baby oil, before placing the hearing aid or earmold into the ear. Petroleum jelly is not recommended for this, since it liquefies as it warms up to body temperature.

3. *High frequency gain reduction*

One solution commonly employed by manufacturers to reduce feedback is to incorporate an electronic high frequency reduction. Some reduction of high frequencies may also be made through acoustic modification, such as the use of a reverse Libby horn.

Horn bores in an earmold or belling the end of the canal in an in-the-ear hearing aid will increase the high frequency gain and will increase the tendency towards acoustic feedback. The reverse of these techniques may be used to reduce high frequency gain. Lengthening the canal portion of the hearing aid will increase the low frequency gain and reduce high frequency gain. The effects of different earmolds, including horns and dampers, is described in Dillon (1985).

In BTE hearing aid fittings, minor modifications to the frequency response may be made by varying the tubing that connects the hearing aid to the earmold. Changing the tubing to a smaller diameter will reduce the height of the receiver peak and shift it to a lower frequency. The acoustic effects of changing from a #18 to #8 tubing, as compared to a standard #13 (1.9 mm ID) tube, are shown in Courtois and Berland (1972). The larger the number of the tubing, the smaller the internal diameter, and vice versa. Similarly, changing to a longer tubing, if possible, will also shift the pri-

	AREA OF FREQUENCY RESPONSE		
	Low Frequencies	***Mid Frequencies***	***High Frequencies***
TECHNIQUE	VENTING	ACOUSTIC DAMPERS	HORN EFFECT (CANAL BELLING)
EFFECT	INCREASED VENTING DECREASES EFFECTIVE LOW FREQUENCY AMPLIFICATION	CHANGING THE ACOUSTIC RESISTANCE (THE DAMPER) VARIES THE MID FREQUENCY PEAKS	CANAL BELLING EMPHASIZES AND BOOSTS THE HIGH FREQUENCIES

Figure 20. Summary of some of the effects of the techniques described in this section on frequency response.

mary peak to a lower frequency. Both of these techniques may help to reduce feedback.

4. *Acoustic damping of receiver peaks*

Acoustic peaks in the frequency response may not appear on 2 cc coupler measurements, but may be created or exacerbated by insertion of the hearing aid into the ear canal. Acoustic modification of the peaks in the receiver response may be made by placing an acoustic damper or small piece of foam in the receiver tubing.

An acoustic damper is an acoustic filter that consists of either a flat mesh screen or sintered metal plug. Dampers may be placed in the receiver tubing of an in-the-ear type of hearing aid, or in the earhook tubing or earmold sound tube when the fitting is a BTE hearing aid. These filters reduce the high frequency gain or "damp" the high frequency peaks, both of which will lessen the tendency towards feedback. The side-effect of using a damper is that the constriction of the receiver tube by the damper material may lead to rapid occlusion by wax, which may require frequent service.

Acoustic dampers primarily affect the mid-frequency resonances that produce feedback in a hearing aid. Acoustic dampers are available with different acoustic resistances. The higher the acoustic resistance, the greater the damping effect.

The specific attenuation of the peak will depend on the type of damper (i.e. foam, fused mesh, sintered plug, star), the acoustic impedance of the damper and the location of the damper in the tubing. It is desirable to verify the specific effect on the hearing aid via probe tube measurements or in a test-box. Damping plugs and material may also be placed in the vent to modify the acoustic feedback pathway through the vent.

Further general information on the effects of different damping materials on various receivers is provided in Dillon (1985) and Knowles (1989b). Figure 20 summarizes the general effects on the frequency response of some of the techniques described in this section.

5. *Use of dual receivers*

Mounting two receivers face-to-face, as illustrated in Figure 21, and driving them in opposite electrical phase will mechanically cancel out many of the vibrations occurring in the receiver case

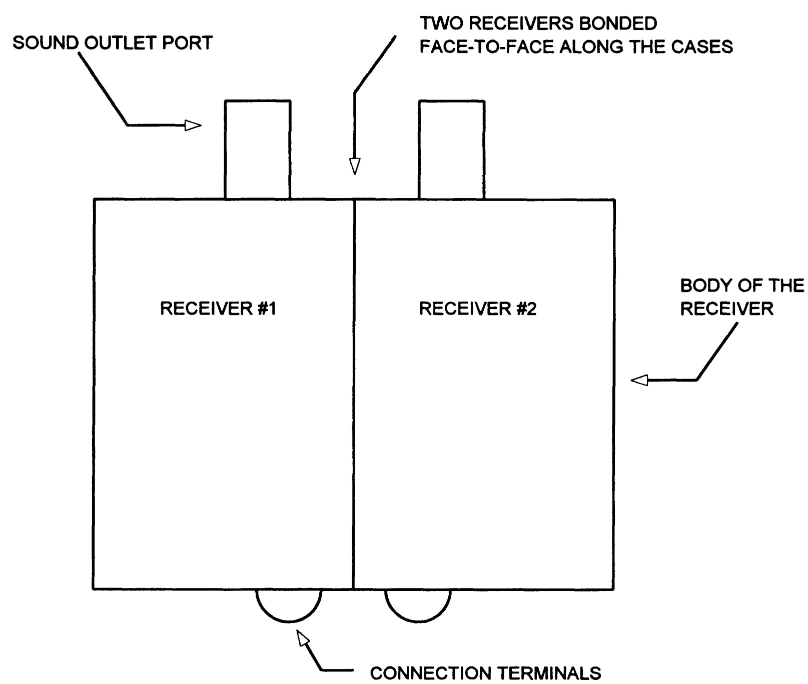

Figure 21. Side view of a dual receiver assembly showing two single receivers mounted face-to-face to cancel and damp case vibrations.

walls. This results in a lower level of direct acoustic radiation from the receiver case, which should lead to a lower level of feedback. A more detailed explanation of dual receivers is provided in Berkowitz (1987). The use of a dual receiver, such as the Knowles Electronics EJ series (Knowles, 1989a), which is equivalent to two receivers mounted face-to-face, is claimed to increase the maximum gain capability by 20 dB (Killion, 1993).

6. *Acoustic notch filtering*

Macrae (1983) has described the use of three types of venting for high-gain hearing aids. These include:

1. A capillary vent; this is a small (0.5 mm) diameter vent.
2. A damped vent; this is a normal (2 mm) diameter vent containing an acoustic damper.
3. A damped cavity vent; this is a venting method in which the vent from the ear connects through an acoustic damper to the inside cavity of the hearing aid (or hollow earmold) and the faceplate of the hearing aid (or earmold) which then has a pinhole vent to the outside. This creates an acoustic notch filter.

Macrae describes the order of the effectiveness of the three in reducing feedback to be that the damped cavity vent is the most effective; the capillary vent is the next most effective; and the damped vent is the least effective.

A more extensive discussion of the measurement of this type of venting is contained in Macrae (1982b). Development of the mathematical basis and measurements of the damped cavity vent has been presented in Macrae (1982a and 1982b).

This type of venting is available as the *Macrae anti-feedback earmold* (Microsonic, 1995).

7. *Modification of the canal tip*

In most cases it is desirable to lengthen the canal portion of the earmold or shell to provide a better seal in the ear canal. This method acts to further separate the receiver outlet from the microphone sound inlet port. Other acoustic modifications that may be useful are lengthening the canal tip, and building up the canal tip or body of the earmold or custom in-the-ear product to create a tighter seal in the ear.

One canal-related modification that may be useful is to order the hearing aid from the factory with an extended receiver tube. In this option, the receiver tube is left so long that it protrudes for perhaps half an inch from the aid and is cut to the desired length during the fitting process. The additional tubing lengthens the feedback pathway between microphone and receiver and may successfully eliminate feedback. By placing the sound output port closer to the eardrum, this often also allows the user to reduce the gain control setting which simultaneously reduces the tendency towards feedback.

In an alternative procedure, Leenen (1995) proposed the use of a CIC hearing aid where the extraction cord, which is normally a short nylon filament extending from the faceplate of the hearing aid, is replaced by a long hollow tube that doubles as an extension of the microphone sound inlet tube. In this way, the microphone inlet is further separated from the receiver and theoretically reduces the susceptibility of the hearing aid to acoustic feedback.

8. *Separation of the microphone and receiver*

Normally the microphone and receiver in BTE or ITE hearing aids are in fixed locations and it is not possible to change the separation between the two in order to reduce feedback. However, there are two extreme cases where separation may be possible.

One is the use of a body-worn hearing aid to provide a large separation between the microphone on the body and the receiver at the ear. The other is to use a CROS (Contra-lateral Routing Of Signal) hearing aid, which provides microphone and receiver separation by placing the microphone on one side of the head and the receiver and sound connection to the ear on the other. Neither of these solutions is generally desirable because of the lack of cosmetic acceptability and because of the difference in sound perception for the wearer when compared to a more conventional ear-level fitting.

CONCLUSION

The audible sound of oscillation caused by acoustic feedback continues to be an annoying, disruptive and undesired fact-of-life for hearing aid wearers. While not all audible artifacts described as *feedback* by the hearing aid wearer are oscillations that can be corrected by modifying the fitting, there are various methods available to suppress or to reduce acoustic feedback through electronic and acoustic modification techniques.

As DSP hearing aids become more prevalent over the next few years, newer algorithms will be developed that will be successful in controlling feedback before it occurs. Until then clinicians must clearly understand the causes of acoustic feedback and be prepared to experiment with different methods of eliminating it in order to produce viable and useful fittings for wearers.

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