Factors Ensuring Consistent Audibility in Pediatric Hearing Aid Fitting

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Abstract
Ensuring consistent audibility is an important objective when fitting hearing aids to children. This article reviews the factors that could affect the audibility of the speech signals to children. These factors range from a precise determination of the child’s hearing loss to an accurate specification of gain in the chosen hearing aids. In addition, hearing aid technology and features such as multichannel processing, directional microphones, and feedback cancellation that could affect the achievement of consistent audibility are reviewed.

Key Words: Audibility, consistency, hearing aid fitting, pediatric

Abbreviations: CR = compression ratio; CT = compression threshold; DI = directivity index; DSL[i/o] = desired sensation level [input-output] formula; DSP = digital signal processing; EAT = equivalent adult threshold; FM = frequency modulation; M = microphone; NAL-NL1 = National Acoustic Laboratories’ nonlinear fitting formula; PC = peak clipping; RECD = real-ear-to-coupler difference; REDD = real-ear-to-dial difference; SNR = signal-to-noise ratio; T = telecoil; VC = volume control; WDRC = wide dynamic range compression
Two fundamental objectives when fitting hearing aids are to ensure audibility of the speech input and verify that sounds are not uncomfortably loud (Skinner, 1980; Popelka and Mason, 1987; Byrne, 1992). A third important objective, especially in the fitting of hearing aids to children, is to ensure consistent audibility and hearing aid performance over time. Consistent audibility is important because young children do not know how to fill in the missing sounds (Nittrouer and Boothroyd, 1990). Consequently, speech sounds that are not heard or are heard only intermittently may not be produced accurately. Production of speech sounds would be delayed (Boothroyd, 1985; Kuk et al, 1999; Higgins et al, 2001). Language markers such as plurality and verb tense may be affected. Furthermore, hearing aids that result in inconsistent audibility (or performance) may place the child in a dangerous predicament in which important warning signals are missed. Although current practice has considered many factors to ensure consistency of audibility and performance, recent advances in digital signal processing (DSP) technology have allowed clinicians to demand even more consistent performance in the hearing aids fitted to children. This article discusses the factors that could affect consistent audibility and indicates the desirable features in today's hearing aids that are more likely to ensure consistent audibility (and performance) for pediatric wearers.

CONSISTENCY STARTS WITH AN ACCURATE MEASUREMENT OF HEARING SENSITIVITY

In order for children to hear all speech inputs consistently, their hearing loss must be measured accurately so that appropriate amplification can be provided. Traditionally, hearing loss is estimated using TDH headphones (-39, -49, or -50) or insert phones. However, because of the significant difference in ear canal volume between an adult and a child under 5 years of age (Bentler, 1989; Feigin et al, 1989), thresholds measured in a young child with a particular transducer reflect not only the degree of hearing loss of the child but also the modification made by the residual ear canal volume with the specific transducer. For example, Figure 1 shows the theoretical difference in hearing levels of a 1-year-old child measured with different transducers and the "true" threshold of the child (equivalent adult threshold [EAT]). The ratio of the residual volume difference between an adult and a 1-year-old child was used to calculate the threshold difference. For the same degree of "true" hearing loss, it is apparent that the thresholds of a 1-year-old child determined with insert phones appear better than those of an adult. On the other hand, the same thresholds measured under headphones or in a sound field are poorer than the average adult at some frequencies. Because amplification is specified by the reported degree of hearing loss, significant gain difference may result depending on how thresholds are obtained. Consequently, hearing loss in a child younger than 5 years of age (on average) should be expressed in EAT to overcome the age and transducer artifacts in threshold measurements (Dillon, 2001) and to ensure accurate gain specification. Currently, only some prescriptive formulae (e.g., desired sensation level [input-output] formula [DSL[i/o]], Seewald, 2000; National Acoustics Laboratories' nonlinear fitting formula [NAL-NL1], Byrne et al, 2001) ask the child's age and the transducer type in calculating gain for the child.

Along the same line, several authors (e.g., Rankovic et al, 1992; Valente et al, 1994) showed significant variations in real-ear sound pressure levels across adult listeners for the same dial reading on an audiometer. This suggests that unless the real-ear sound pressure level at threshold is measured, thresholds determined with any transducers are likely to show real-ear variability across individuals, especially in children (Lewis and Stelmachowicz, 1993). An error in threshold estimate will lead to a wrong assignment of gain.
Determining the real-ear sound pressure level at threshold is not an error-free task. Depth of probe insertion, location of the probe tube, and location and azimuth of the loudspeaker can lead to variable results (Dirks and Kincaid, 1987; Valente et al, 1990; Rankovic et al, 1992; Dirks et al, 1996; Zelisko et al, 1996). Furthermore, the use of a real-ear measure in children may be especially challenging when cooperation and fatigue would not only increase the variability of the measured output but would also limit the amount of information (e.g., number of frequencies) that one can obtain. For that matter, group real-ear-to-dial difference (REDD) correction factors for different age groups and different transducers (Scollie et al, 1998) are welcome tools that may improve the accuracy of threshold estimation while remaining clinically efficient. The use of individual real-ear-to-coupler difference (RECD) or the use of group RECD (Zelisko et al, 1992; Westwood and Bamford, 1995; Seewald et al, 1999) also increases the ease with which one can select the right coupler response from a selection of hearing aids that may yield the desired insertion gain. Obviously, group REDD and group RECD represent average data. Individual variations in REDD and RECD from the group data would occur and minimize its predictive accuracy.

On the other hand, if one recognizes that the source of variability arises from factors such as the difference in residual volume, impedance, etc., one may avoid such variability through the use of in situ (i.e., hearing aids in ear) threshold measurement. During this procedure, the child wears the actual hearing aids with which he/she will be fitted (Ludvigsen and Topholm, 1997). Acoustic signals are generated from the hearing aids for threshold measurement. The residual volume, venting, and impedance of the child's ear canal will modify the magnitude of the acoustic output from the hearing aids. The electrical output at which real-ear threshold is indicated will be noted by the hearing aids. The necessary electrical output to reach a required sensation level (and insertion gain) can be estimated by the fitting software automatically. As few as three to four signals are necessary to characterize the degree and configuration of the hearing loss.

In situ threshold technique assumes that the child is able to perform a behavioral threshold task. Although it does not reduce the variability inherent in behavioral threshold measures, it has the advantage of accurately specifying the "real-ear" threshold without the requirement of a probe microphone system, that is, convenience. Because the thresholds are determined with the child's hearing aids in place, any effects of volume, leakage, and so on are reflected in the magnitude of the thresholds. There is no need to determine or apply correction factors such as REDD and RECD to predict the real-ear gain from the coupler-gain curve. One can determine these values from the fitting software for the hearing aids. Indeed, one would be able to estimate the amount of gain much more accurately than with actual real-ear insertion gain measures because no measurement errors, which are inherent in probe microphone measures, may be introduced. For the average child, the need to verify real-ear gain with probe microphone measures may be reduced. However, it does not replace the importance of probe microphone measures. Indices such as the real-ear saturated response or discomfort level can be determined only with real-ear measurement. A limitation of current in situ threshold measurement technique is that it is available only in a few digital hearing aids. Furthermore, the results obtained with one brand of hearing aid cannot be transferred to other hearing aid brands. It is foreseeable that this technique may be adopted by other digital hearing aids in the future.

**CONSISTENCY IS ENSURING THE BEST FORM OF AMPLIFICATION**

The first factor to ensure consistent input for the pediatric patient with bilateral hearing loss is the consistent use of binaural hearing aids. Auditory developments such as localization, release of masking in noise, etc. are highly dependent on the availability of binaural input (Kock, 1950; Haggard and Hall, 1982; Yost, 1994; Hood, 1997). Furthermore, children learn from hearing sounds, both speech and nonspeech, from their auditory environments. Aiding a child monaurally in the presence of a bilateral hearing loss could delay the auditory development and speech and language skills of the child. It may affect the child's social and academic progress. Recent findings on the deprivation effect should further warn against monaural hearing aid use in bilateral hearing loss (Silman et al., 1984, 1992; Silverman and Silman, 1990). Furthermore, the safety of the child may be jeopardized when warning signals presented on the unaided side are missed.

In addition to binaural amplification, the type of signal processing may also be impor-
tant. Earlier forms of amplification for children were primarily linear amplification with either peak clipping (PC) or compression limiting (CL) to limit the output. Although much is known about the relative merits of different forms of amplification, almost half of the hearing aids dispensed to children, especially to those with a severe-to-profound degree of hearing loss, are still linear hearing aids with PC or CL (Tharpe et al, 2001).

The advantage, as well as the disadvantage, of linear hearing aids is that equal gain is provided at all input levels until saturation. A fixed gain maintains the relative intensity contrast of the input signals and preserves the intensity cues within the speech signals. Such cues could be important in speech understanding, especially with increasing hearing loss (Van Tasell et al, 1987, 1992; Van Tasell, 1993). On the other hand, a child seldom stays in listening environments that have the same range of intensity fluctuation. From a soft whisper that a mother made to her child, or someone calling the child at a distance, to parents playing with their child on their laps, a child will necessarily hear sounds at different input levels. Indeed, Stelmachowicz et al (1993) illustrated the problem by showing the variation in input levels at children's ears when their parents vocalized at different distances. At a fixed gain setting, as in the case of linear hearing aids that are fitted to optimize amplification for a medium or conversational level input, the child may be overamplified in some instances and underamplified in others. Consistent audibility and listening comfort may be jeopardized. The problem is especially aggravated in the case of a severe-to-profound hearing loss in which the necessary high gain on the hearing aids may result in a constantly high output that saturates the hearing aids at medium- and high-input levels. It is known that constant exposure to a high output could lead to additional hearing loss (Macrae, 1991, 1993, 1995). A way to vary gain for different input levels (and sometimes the frequency response) may be necessary.

Gain variation can be achieved through the use of volume controls (VCs) that are available on all linear hearing aids. Another option is the use of multiple memories that feature a different frequency-gain response designed for a specific listening situation. The rationale behind the use of VCs or multiple memories is that the fixed processing within linear hearing aids, although optimal for a conversational input, may not be optimal for processing in other environments. Although the validity of this rationale is contingent on the processing offered by the specific hearing aids, the assumption that the VC and the remote control or memory buttons can be consistently and successfully used by children may not be valid. Some older children may find the use of a remote control (or VC) acceptable; younger children or infants would have neither the cognition nor the physical prowess to change the VC or memory button. Indeed, the presence of a VC or memory button may actually increase the variability of the intended gain use because the child may accidentally change the intended setting. It is unreasonable to expect the caretaker to ensure proper gain/memory setting consistently all of the time. The use of a VC or program button to alter gain (and frequency response) setting on hearing aids for the fixed gain dilemma may not be a practical solution for young children.

A possible and practical solution for consistent audibility at more (if not all) input levels is the use of wide dynamic range compression (WDRC) hearing aids. By design, such hearing aids can provide more gain for low-input levels and less gain for high-input levels than linear hearing aids when both are matched in gain for a medium-input level. This is shown in Figure 2, in which a WDRC hearing aid and a linear hearing aid are matched in output at a medium input level (60 dB SPL).

![Figure 2](image)

Figure 2 Hypothetical difference in output between a linear hearing aid and a wide dynamic compression range hearing aid when both are set to yield the same output at a medium input level (60 dB SPL).
CONSISTENCY IS ENSURING OPTIMUM GAIN AT ALL INPUT LEVELS

There is no clear-cut recommendation on the precise gain that should be prescribed for children to ensure consistent audibility. In the adult literature, there are no studies to suggest that the results of one prescriptive formula are superior to others. Such studies are even more difficult to conduct in children because infants and very young children may not be able to provide reliable preference judgments, and the preference data from older children may be (at least partially) biased by the hearing aids that they were fitted with at a younger age (e.g., Snik and Hombergen, 1993; Snik et al., 1995; Ching et al., 1997). Consequently, one may need to formulate the optimal gain for children through a consideration of available psychophysical and developmental data. Needless to say, gain assignment for young children would need to start with a prescriptive approach. The results of the fitting should be closely monitored and fine-tuned as the child becomes older.

Need for Level-Dependent Corrections

One needs to be clear on how the desired real-ear gain should be prescribed prior to the discussion of optimal real-ear gain for varying input levels. Earlier it was mentioned that a child’s hearing thresholds should be specified in EATs. On the other hand, gain assignment based on the EATs assumes an adult-size ear canal volume. This is not a valid assumption, especially for children under 5 years of age, in whom the ear canal shows the most growth (Kruger, 1987; Bentler, 1991). Thus, for very young children, the assigned gain based on EAT must be modified to reflect the difference in ear canal volume between the child and the adult. Unless real-ear thresholds are used to specify gain and real-ear measures are used to verify in situ gain, errors would occur when group correction factors are used to account for the adult-child ear canal volume difference, especially in the case of nonlinear hearing aid fitting and for very young children. For a more detailed discussion of this issue, readers are referred to Marcoux and Hansen (2002). A numeric example will best illustrate the difficulty when using group corrections with a nonlinear hearing aid fitting.

Let us assume that a 1-year-old child has a hypothetical threshold of 60 dB HL measured with an insert earphone at a particular frequency. Let us also assume that the EAT for the child at this frequency is 70 dB HL. This suggests that an adult-child ear canal volume correction of 10 dB has been applied. Let us examine the potential gain difference between using the measured thresholds (with the inserts) and the EAT for linear and nonlinear hearing aids.

Linear Hearing Aid

If one assumes that a half-gain rule provides optimal gain for the child, one would assign 30 dB of real-ear gain at the particular frequency for all input levels (low, medium, and high) when the hearing loss is measured at 60 dB HL. For the EAT of 70 dB HL, 35 dB of real-ear gain would have been assigned if the loss were measured in an adult. Because of the age of the child, 10 dB (from the adult-child ear canal volume difference) will need to be subtracted to yield a real-ear gain target of 25 dB. A difference of 5 dB of gain is noted across all input levels between the two gain targets (i.e., 30 vs 25 dB).

Nonlinear Hearing Aid

If one assumes a full-gain rule for low-input level sounds, a half-gain rule for medium-input level sounds, and a quarter-gain rule for high-input level sounds, one would assign 60 dB of gain for low-input levels, 30 dB of gain for medium-input levels, and 15 dB of gain for high-input levels when the hearing loss was measured at 60 dB HL. For the EAT of 70 dB HL, and applying the same 10 dB adult-child correction to the prescribed gain, one would obtain a gain target of 60 dB (70 – 10) for the low-input levels, 25 dB (35 – 10) for the medium-input levels, and 7.5 dB (17.5 – 10) for the high-input levels. Although there is no gain difference between the two gain targets for the low-input levels, differences of 5 and 7.5 dB are noted for the medium- and high-input levels, respectively. Figure 3 illustrates the theoretical difference between a linear and a nonlinear hearing aid.

Such gain differences and, indeed, the whole issue for the need of EAT and adult-child correction may be eliminated if thresholds and gain are measured in situ using real-ear measurement systems. On the other hand, given the
variability in real-ear measures and the limited cooperation from very young children, use of group corrections such as REDD and RECD (group as well as individual) are necessary but not error-free shortcuts. This is because these corrections are based on a level-independent (i.e., linear) assumption. That is, the amount of correction is the same regardless of input levels. The use of average corrections with nonlinear hearing aids may be questionable because of gain changes with varying input levels. Level-dependent corrections may be necessary for nonlinear hearing aid use.

**Gain for Low-Level Input**

As was indicated earlier, a child's ability to fill in missing segments is poorer than that of an adult (Nittrouer and Boothroyd, 1990; Nittrouer, 1996). Much of their learning is not from structured, formal settings but from play activities in which sound sources vary considerably in their azimuth, input level, and spectral content. This makes the requirement “to hear sounds at a distance” not an unreasonable one for a child. Any prescriptive approach would need to ensure audibility of all meaningful sounds. To do this, one needs to provide more gain to children for low-input levels than would be necessary for adults. Psychophysical findings also support that speculation. Nozza and colleagues (1991a) showed that normal-hearing infants need as much as 26 dB more sound pressure level than adults to perform comparably on a discrimination test. By providing more gain for low-input levels, children will be given the opportunity to hear softer sounds. Providing extra gain for low-intensity sounds on WDRC hearing aids may have a lower risk of overamplification (than linear hearing aids) because such hearing aids decrease their gain with increasing input level.

There are no clear-cut answers on how much gain for low-input sounds one should provide a child. However, one may speculate on such a requirement by making certain assumptions on what may be desirable for a young child to hear. If one assumes that the lowest level that the child needs to hear occurs at 0 dB HL, the desired gain for low-input level sounds should be equal to the degree of hearing loss of the child. On the other hand, if the criterion is to allow the child to hear low-input level speech (and not the softest sound), then less gain will be required. Considering that the level of soft conversational speech occurs at around 20 dB HL (Ross et al, 1991), a gain value that is equal to the degree of hearing loss less 20 dB should be sufficient to bring the lower range of normal speech to the child’s hearing threshold. The difficulty with achieving such high gain value, especially for a severe-to-profound loss, is the potential risk of feedback, which increases as gain for low-level sounds increases (Kuk, 1999). With the advent of DSP and the inclusion of various types of feedback management algorithms in many digital hearing aids, significantly greater gain for low-input level sounds can be realized in many DSP hearing aids today.

A good way to ensure that the child receives optimal gain for low-input sounds is to determine the aided soundfield threshold of the child with the chosen nonlinear hearing aids. Because the aided threshold for a nonlinear hearing aid (without VC adjustment allowed) represents the softest sound that the wearer hears (Kuk, 2000), the aided threshold could indicate if any of the speech spectrum (or any sounds) is audible to the child. For example, an aided threshold at 20 dB HL would suggest that the child has the potential to hear the softest part of normal speech. A low aided threshold ensures audibility of low-intensity speech and can result in more consistent speech perception/recognition of low-level speech (ignoring potential issues with dead regions). Indeed, Gabbard and colleagues (2000) demonstrated an increase in speech recognition score (over subjects’ own linear hearing aids) at a low-input level (50 dB SPL) in children with a severe-to-profound hearing loss. Obviously, the determination of the aided threshold requires that the child is cognitively able to perform such a task. With early amplification (and sound conditioning), Ramos and colleagues (1999) were able to fit infants as young as 5 weeks and reliably performed visual reinforcement audiometry on the same infants at 5 to 9 months. Necessary precautions to minimize
Gain for Medium-Level Input

The “correct” gain target for medium-level input is difficult to define also. After all, there are many linear and nonlinear prescriptive fitting formulae, each recommending different amounts of gain. Figure 4 shows the difference in target gain between the DSL[i/o] (Cornelisse et al, 1994; Seewald, 2000) formula and the NAL-NL1 formula (Dillon, 1999; Byrne et al, 2001) for a relatively flat hearing loss at three input levels (40, 65, and 90 dB SPL). It is easily seen that gain for the high-input level is quite similar between these two prescriptive targets. However, gain for medium- and low-input levels is quite different between the two prescriptions.

Despite notable differences in gain prescription across formulae, most research conducted on adult subjects with linear amplification concluded that as long as audibility for low-input level sounds and comfort for high-input level sounds are ensured, most linear prescriptive formulae do not differ much in their real-life efficacy (Horwitz et al, 1991; van Buuren et al, 1995). In other words, “matching” precisely the prescriptive target for medium-level input may not be as critical as long as one can ensure audibility of the lowest-input sounds and comfort for the highest-input sounds. Because gain targets that attempt to equalize loudness across frequency yield the highest speech intelligibility (Byrne, 1996), a reasonable approach is to select a prescriptive target that ensures equal comfortable loudness across frequencies for medium-input levels and a gain target that ensures audibility and listening comfort for low- and high-input levels, respectively. Although there is no evidence to suggest that one nonlinear prescription is superior to the others, investigation is under way at the NAL and the University of Western Ontario that compares the relative efficacy between the NAL-NL1 and DSL[i/o] fittings in children (Ching et al, 2001).

Target Matching

It is a common practice for many clinicians to measure the real-ear (or the coupler) output of hearing aids and match it against a prescriptive gain target to ensure that the child receives the desired amount of gain. Although this practice may offer a convenient way of ensuring adequate gain/output when fitting linear hearing aids, its use may be questionable when verifying the fittings of multichannel non-
linear hearing aids. The number of channels, the time constants (attack and release time), and so on modify the output of the hearing aids (Kuk and Ludvigsen, 1999). In addition, the choice of stimuli to evaluate nonlinear hearing aids may also result in variable output unless special precautions are taken to minimize such occurrences. Unfortunately, most prescriptive targets have not considered the effects of these variables on the output of the hearing aids. Thus, matching the output of nonlinear hearing aids to these generic gain targets may compromise the fitting rather than ensure a successful fit. For example, Stelmachowicz and colleagues (1998) showed that the preferred gain selected by adult experienced hearing aid wearers was closer to the manufacturer's target than to a generic prescriptive target.

In selecting a prescriptive formula, it is important to know not only how much gain is prescribed but also its underlying principles and how much consideration has been made relative to existing technology. For example, the NAL-NL1 formula considers the effect of number of channels (up to four only) in its gain formulation, whereas no other formulae have yet to consider this effect. Thus, the criterion to consider in a fitting is not selecting the "right" prescriptive formula but rather the acceptability of the aided results (e.g., aided thresholds) from the use of the prescriptive gain formula. Because the aided threshold reflects the lowest-input level that the listener hears at a specified setting, a lower aided threshold could be viewed more favorably than a higher aided threshold. This may ensure more consistent amplification of low-input level sounds. However, this simple conclusion is based on the assumption of nonlinear WDRC hearing aids instead of linear hearing aids and that children rather than adults are considered.

**CONSISTENCY IS PRESERVING AS MANY INPUT CUES AS POSSIBLE**

Although WDRC hearing aids may ensure better audibility across listening situations than linear hearing aids, the speed at which gain changes (i.e., attack and release times) merits further consideration. Typically, a fast-acting WDRC hearing aid (attack time less than 5–10 msec, release time less than 100 msec) is designed to have gain follow the rapid intensity fluctuation of the input stimulus. By doing so, this action could lead to a reduction of the intensity contrasts within the acoustic stimulus. By taking a longer time to complete this gain change, a slow-acting WDRC hearing aid (attack time greater than 50 msec, release time greater than 1 or 2 sec) preserves the intensity contrasts of the input signal better while adjusting the overall gain appropriate for the environment. Indeed, by extending the attack and release times of a WDRC hearing aid, one effectively reduces the static CR of the WDRC hearing aid in the real world. It becomes more linear than the static CR would suggest (Stone and Moore, 1992; Verschuure et al, 1996; Fortune, 1997; Kuk, 1998a). Figure 5 shows the waveform difference of a speech segment processed by a WDRC hearing aid with a 200-msec release time (lower curve) and one with a 2-sec release time (upper curve). Note that the peaks and valleys are more clearly demarcated in the waveform processed with the longer release time. Neuman and colleagues (1995) and Hansen (2002) showed that adult hearing-impaired subjects preferred a longer release time than a shorter release time in single-channel and multichannel WDRC hearing aids, respectively.

The temporal waveform provides an important cue for speech perception because the prosodic information and suprasegmental cues are embedded within the temporal waveform (Stone and Moore, 1992; Van Tasell et al, 1992; Drullman et al, 1994; Verschuure et al, 1996; Fortune, 1997). Alteration of the waveform may not only change its audibility and intelligibility, it may also alter the hidden meaning and emotional intent of the utterance. Furthermore, the reliance of adult hearing-impaired subjects on the temporal envelope for speech identification increases as the listeners’ hearing loss extends beyond a moderate degree (Van Tasell et al, 1987). It is not unreasonable to expect the same or greater reliance of this cue in children because they, especially the younger ones, require large stimulus difference for proper speech identification (e.g., Nozza et al, 1991a). At least in theory, the use of fast-acting compression may make...
it more difficult for some younger children to distinguish between meaningful and nonmeaningful signals. For example, during the pauses of speech, the fast attack and release times of WDRC hearing aids would amplify the low-level ambient noise of the room. This could make it more difficult for the young hearing-impaired children to discriminate between meaningful “noise-like” speech cues, for example, fricatives, and nonmeaningful ambient noise. The use of slow-acting WDRC hearing aids could minimize such an occurrence and provide a more contrastive input for the child. Dillon (2001) also suggested the use of slow-acting WDRC or adaptive release time in WDRC hearing aids to ensure consistent audibility, signal contrast, and comfort. Clearly, more research is necessary to study the effect of attack and release times in WDRC hearing aids for children.

**CONSISTENCY IS ENSURING SPECIFICITY OF PROCESSING**

Not all children have the same configuration of hearing loss. How well one shapes the frequency response of hearing aids to match the child’s audiometric configuration may affect how consistently a child hears (or does not hear) particular sounds.

One of the features of current compression hearing aids that increases the flexibility of frequency response adjustment is the availability of multiple processing channels. Although single-channel hearing aids offer some flexibility in frequency response adjustment through the use of different microphones, additional resistors, capacitors, etc., such flexibility is generally restricted. Additionally, because there is only one processing channel in this type of hearing aid, any gain change in one frequency region may also affect gain at other frequency regions. This can be illustrated in Figure 6A, which shows a hypothetical situation in which a single-channel system is required to reduce gain in a restricted low-frequency region where noise is detected. To do that, gain at other frequencies would be lowered as well. On the other hand, Figure 6B shows an example of a two-channel system. Assuming complete channel independence, gain reduction in the low-frequency region is confined to the low frequency, sparing the higher frequencies from gain reduction.

As one may imagine, as the number of independent channels in compression hearing aids increases, the specificity of gain change increases. This means that any unnecessary or unintended gain reduction (or increase) would occur less frequently than it would in hearing aids with fewer channels. This may ensure that the desired inputs are consistently amplified while the undesirable signals are minimized. Situations in which such specificity may be especially helpful would include hearing aids that have noise reduction algorithms or that have mechanisms for controlling feedback.

The use of WDRC hearing aids with multiple channels may improve the speech production skills of some children with a hearing impairment. This is because a multichannel device may allow children to monitor their own vocal production better than a single-channel device. Because of the radiation characteristics of the vocal apparatus and the effect of distance, speech produced by a speaker at a normal level would measure 10 to 15 dB higher at the level of the ear than at a conversational distance of 1 meter (Dunn and Farnsworth, 1939; Cornelisse et al, 1991). This is shown in Figure 7, which shows the increase primarily in the low and mid-frequencies. If the child wears single-channel linear hearing aids, and assuming that the VC is not adjusted, the higher-input signal would result in a higher output. This additional output in the low frequencies could mask the higher frequencies and render them inaudible (upward spread of masking), or it may saturate the hearing aid to result in distortion. Even if the child is able to lower the VC to achieve a comfortable loudness, such an effect will reduce the overall gain across all frequencies and input levels. In all of these scenarios, the higher-frequency sounds may not be audible to the child. If the
Figure 7 Difference in the speech spectrum level as the speaker's voice was recorded at a typical conversational distance of 1 meter and at the ear level of the speaker.

If a child wears single-channel WDRC hearing aids, the high input would cause gain reduction across all frequencies to render the high frequencies inaudible as well. It would be difficult for children to improve their speech production skills if they cannot hear/monitor their own production consistently.

In a two-channel system, the high-input low-frequency signals at the microphone would activate compression in the low-frequency channel and reduce its low-frequency output. This may spare gain in the high frequencies and preserve audibility of the high-frequency sounds. This may allow the child to hear/monitor his/her own voice more consistently. Such consistent input may improve a child's speech production skills. In two separate studies with a multi-channel digital hearing aid, many of the 8- to 12-year-old subjects and/or their parents reported an improvement in the children's speech production ability. More consistent articulation, less nasality, and more appropriate voice level were reported (Kuk et al, 1999; Gabbard and O'Grady, 2000). Direct studies examining this potential are needed.

Although multichannel hearing aids have the potential to offer more specific processing, not all multichannel products offer the same degree of specificity. Commercial multichannel hearing aids differ in the number of channels, bandwidth of each channel, slope of each filter channel, and degree of independence across channels. This could have a bearing on the specificity of the processing. Furthermore, as the number of processing channels increases, the potential of spectral and temporal smearing also increases (Baer and Moore, 1994; Drullman et al, 1994; Boothroyd et al, 1996). This has the effect of diminishing the intensity and spectral contrast of the input signals and may lead to a potential decrease in sound quality and intelligibility of the speech input (e.g., Bustamante and Braida, 1987; Yund and Buckles, 1995). In choosing a multichannel compression hearing aid, it is important to understand how the hearing aid is designed to achieve maximum specificity while minimizing potential artifacts.

**CONSISTENCY IS ENSURING A FAVORABLE SIGNAL-TO-NOISE RATIO WITH MINIMAL SENSITIVITY LOSS**

**Directional Microphones**

Children need a higher signal-to-noise ratio (SNR) in noisy backgrounds to reach the same level of performance as adults. This is demonstrated in the works of Nozza and his colleagues (Nozza et al, 1991b), in which they showed that normal-hearing infants required as much as 7 dB higher SNR to perform similarly as adults in a discrimination task. Boothroyd (1997) also showed that preschool children required an average of 3 dB higher SNR than adults for comparable recognition performance in noise.

The use of hearing aids with directional microphones and the use of direct audio input devices such as frequency modulation (FM) systems have been advocated to achieve a higher SNR. The use of directional microphones in adults has received ample support. However, the number of references supporting the use of directional microphones in children is more limited (Hawkins, 1984; Gravel et al, 1999; Kuk et al, 1999). This is quite the contrary in the case of FM use. One unsettling issue is that both FM and directional microphones could affect the audibility of some input that children receive through their hearing aids.

A directional microphone is based on the principle of selective attenuation of microphone sensitivity to sound sources arriving from different azimuths (Valente et al, 1999; Csermak, 2000; Kuk et al, 2000). Its principle assumes that the signal at 0 degrees azimuth is the primary signal of interest, whereas signals presented anywhere else are noise or undesirable in nature. This assumption has been known to be invalid in some communication situations for adults (e.g., taxi drivers, who need to hear passengers from the back). Children move about freely by nature and learn as much (maybe even more) during play activities as in formal learning situations. Thus, the assumption of a fixed signal source is more likely to be challenged. In situa-
tions for which the assumptions for a directional microphone are no longer valid, that is, speech from the sides and back, noise (or undesirable signals) from the front, the use of a directional microphone could negatively impact speech understanding. Indeed, the more effective a directional microphone (higher directivity index [DI]), the more negative impact it may have on audibility (Gravel et al., 1999). Lee and colleagues (1998) showed a 24 percent decrease in speech intelligibility for soft speech (50 dB SPL) between a directional microphone and an omnidirectional microphone of the same hearing aid when speech stimuli were presented from the back of adult hearing-impaired subjects. This limitation of directional microphones could affect the audibility of the inputs received by children in at least some listening situations. In addition to missing potential learning opportunities, it could also place children at risk of not hearing warning signals while at play. This may be one reason why a directional microphone is recommended to children only with caution (Pediatric Working Group, 1996).

There are additional limitations to the use of a directional microphone, such as higher susceptibility to wind noise and circuit noise level. The readers are referred to Kuk and colleagues (2000) for a detailed description of some of these limitations. These limitations prompted some manufacturers to design directional hearing aids that allow the wearer to manually switch between an omnidirectional and a directional microphone. In a quiet environment or at play, one can instruct the child to leave the microphone in the omnidirectional mode so that wind noise and circuit noise are minimized while the sensitivity of the microphone is maximized, whereas in structured learning situations in which the noise can be confined to the sides and back, the aid can be switched to a directional mode. Although this is a plausible solution, its effectiveness is dependent on the child (or the care taker) in making the correct microphone switch at the right time. Although older children may have the cognitive and physical ability to perform such a task, it is unreasonable to expect young children and infants to be able to switch. Some mechanisms are necessary to ensure consistent audibility of the input signal at the best SNR.

Hearing aids with adaptive beam forming are currently available that change their directionality depending on the location of the noise source. These hearing aids achieve the adaptive directional patterns by integrating signals from two (dual) omnidirectional microphones using DSP. However, because the resulting DI of a dual-microphone system relies on the operation of two perfectly matched omnidirectional microphones, any drifting in performance relative to the sensitivity and/or phase characteristics of either microphone could result in an unintended polar pattern. This not only decreases the effectiveness of the directional microphone, it may also result in a polar pattern that reduces the sensitivity of the microphone to sounds from the front but maximizes its sensitivity to sounds from the back (e.g., Kuk et al., 2000). Proper microphone matching (both sensitivity and phase) is necessary in a dual-microphone system to ensure a consistently favorable SNR.

The advantage of adaptive directionality over a fixed directional microphone is a consistent SNR improvement (over an omnidirectional microphone) regardless of the noise location. On the other hand, current directional microphone systems are not capable of distinguishing between speech and noise. Speech (or any desirable signal) presented from the sides and back would still be attenuated. Directional microphones that are coupled to WDRC hearing aids that provide extra gain for low-input sounds (such as a low CT) may partially overcome such a problem and render the output audible to the wearers even when the desirable signal is presented from the back. Lee and colleagues (1998) showed that a digital directional hearing aid with a low CT yielded a higher speech recognition score when speech at 50 dB SPL was presented from the back than another WDRC hearing aid with a higher CT even in its omnidirectional mode (44% vs 37%, p < .05).

The rate at which the polar pattern changes from an omnidirectional mode to various directional patterns in response to the change in signal location may be critical in ensuring audibility for desirable signals presented to the sides and back. If the polar pattern changes too quickly, the wearer may not even notice that a desirable signal has been presented. A slower rate of polar pattern change may allow wearers time to turn their heads accordingly to compensate for the sensitivity loss. These issues again reinforce the Pediatric Working Group's (1996) position for the need to carefully consider how a directional microphone may affect a child prior to its recommendation. Current digital hearing aids that allow a choice between omnidirectional and adaptive directionality may make the choice easier for clinicians.
Figure 8 Demonstration of transparency: a frequency modulation (FM) receiver coupled to a hearing aid yields the same frequency-gain output from the hearing aid as when the hearing aid is tested by itself (A) and when the stimuli were presented to the hearing aid microphone in the FM + hearing aid mode (B), the FM microphone in the FM mode (C), and the FM microphone in the FM + hearing aid mode (D).

Direct Audio Input

An FM system has been used to further improve the SNR of the learning environments, especially for school-aged children (Hawkins, 1984). For children to receive a consistent input in school and at home, the output from the FM receiver is typically coupled to the child's hearing aids. Obviously, it is based on the assumption that the introduction of the FM to the hearing aid system would not alter the frequency-response and gain characteristics of the hearing aids.

An FM system that does not alter the frequency-gain characteristics of the hearing aid is said to be transparent. That is, if one presents a broadband signal to a hearing aid that is coupled to a transparent FM system, one would measure the same output from the hearing aid regardless of where the input is presented (i.e., to the hearing aid microphone or FM microphone) as long as it does not saturate either the FM or the hearing aid. Figure 8 illustrates the concept of transparency with the coupler frequency-gain curves of a digital hearing aid coupled to a wireless FM receiver. A speech-shaped noise from 50 to 80 dB SPL was presented in 10 steps to either the hearing aid microphone (see Fig. 8, A and B) or the FM microphone (see Fig. 8, C and D). The FM/hearing aid settings are hearing aid alone (A), FM alone (C), and FM + hearing aid (B and D). Note that for the same input levels, the same frequency-gain curve was obtained until 80 dB SPL, where the input compressor within the FM transmitter may have lowered the overall gain to prevent saturation distortion.

One would imagine that all FM systems should be transparent to the hearing aids to which they are coupled. Unfortunately, this is not the case. Various authors (e.g., Hawkins and Schum, 1985; Thibodeau, 1990) showed significant changes in the final frequency response/output of some hearing aids when coupled to some FM systems. Consequently, to ensure that children with a hearing impairment receive consistent inputs, one needs to ensure that the FM system is transparent to the hearing aids. Newer FM systems are designed to be transparent. Older FM systems may require significant adjustments on the FM by the clinicians in order that the output from the hearing aid with the FM is similar to the hearing aid alone condition.
A related issue is the use of the inductive coil (or T-coil) on the hearing aid. This could be useful in some schools in which an inductive loop is used for classroom instruction for children with a hearing impairment. In addition, many children use the T-coil on the telephone. The assumption is that the same frequency response characteristics are achieved in the "M" (microphone) and the "T" (telecoil) modes. Unfortunately, in almost all conventional analog hearing aids, the sensitivity of the microphone response is different from the telecoil response (Downing, 2000). Figure 9 shows the individual difference curves between the T-coil and the microphone for 24 hearing aids when the VC was set at the reference test gain position. A frequency gain difference of zero should be observed (across the middle of the graph) if the T-coil sensitivity was the same as the microphone. As noted, of the 24 hearing aids tested, 6 of the T-coils had a higher sensitivity in the telecoil mode, whereas 18 had a higher sensitivity in the microphone mode. This means that when a child switches a hearing aid from the M position to the T position, a significant loss in low- and high-frequency responses may result. Although a child can increase the volume of the hearing aid to partially offset the loss in loudness, loss of frequency-specific information is inevitable. In situations in which the child may not have the cognition to increase the volume, or if the teachers or the caretakers who are responsible to switch the hearing aid from M to T are not aware of such a frequency difference, the child would receive inconsistent input between the two transducer modes. Care must be taken to ensure the same sensitivity between the T mode and the M mode.

With analog hearing aids, the wearer can compensate for the loudness difference only by increasing the VC. With some multimemory programmable hearing aids, the clinician can designate another memory for the T function in which increased sensitivity (and a different frequency response) may be programmed in the T-coil to match the loss of microphone sensitivity. With some digital hearing aids, an automatic compensation is made when the wearer switches from M to T. These alternatives are necessary to ensure the consistency of the auditory input.

**CONSISTENCY IS ENSURING THE ABILITY TO USE THE PRESCRIBED GAIN SUCCESSFULLY**

Although one may be able to identify children at a very young age and provide them with the appropriate hearing aids, such efforts may be futile if these children cannot wear the hearing aids in real life because of problems associated with feedback. This is a problem for young children, whose ears continue to grow until they are 8 to 12 years of age (Riedner, 1978). The problem may be more serious for infants who are less than 20 months of age, in whom rapid growth in the size of the ear canal is observed (Kruger, 1987; Feigin et al, 1989). This suggests that even when new earmolds are made frequently, feedback may prevent a child from using hearing aids consistently at their prescribed gain setting.

Another reason for hearing aid feedback in infants and young children is the likelihood of their heads being close to reflective surfaces. Such is the case when they are lying in bed or when being held by their caretakers. The close proximity between the reflective surfaces and the children's ears exacerbates any imperfections (or leakage) of the earmolds and can lead to feedback.

Most people would lower the VC setting (i.e., reduce gain) on their hearing aids to stop feedback. In single-channel linear hearing aids, lowering the VC eliminates feedback at the expense of reducing gain at all frequency regions, including those in which feedback does not occur. Furthermore, gain at all input levels is lowered. This action overrides the clinicians' efforts in achieving the desirable frequency-gain characteristic and may significantly compromise audibility and intelligibility.

Kuk and colleagues (2002) described some of the approaches that have been used to manage feedback and the associated challenges with
each approach. These approaches vary in their levels of technical complexity as well as the associated cost. Although not every child would require the most sophisticated system (many can simply use a well-fit earmold) to prevent feedback, it is likely that the youngest children will have the greatest need for the more sophisticated systems because of their growing ears and closer proximity to reflective surfaces. A feedback management system for this age group must allow sufficient gain for all input levels to ensure audibility of even the softest speech. In addition, it must be able to accommodate the gradual changes in the feedback path as the child (and ear canal) grows. In addition, it must be able to respond quickly to the changes in the feedback path, for example, when the child is put to bed or is held by the parents. These requirements would suggest that the ideal feedback management system for the youngest children must be adaptive in nature. Obviously, a well-fit earmold is still critical.

An adaptive feedback cancellation system monitors the feedback path continuously. When a feedback signal is first detected, it generates a feedback cancellation signal that is identical in frequency and amplitude but opposite in phase to the feedback frequency. This cancellation signal cancels the frequency causing the feedback at the input. Because the system continuously samples the feedback path and updates its characteristics, changes in the feedback path (e.g., the child's ear growing in size or the child being held to the parents' chest) could be immediately compensated. Thus, even if the earmold may become loose following growth of the ear canal, it may not be necessary to immediately replace the earmold. However, when the growth of the ear canal exceeds the potential gain offered by the feedback mechanism, feedback would still occur. A new and tighter earmold will need to be ordered. In general, one may expect an average increase of 10-dB usable gain from an active feedback cancellation system in today's commercial systems (Bisgaard and Dyrlund, 1991; Kates, 1991; Engebretson et al, 1993; Ludvigsen and Kuk, 2001).

**CONSISTENCY IS ENSURING MINIMAL ARTIFACTS UNDER ALL CONDITIONS**

**Current Drain**

Children with a severe-to-profound hearing loss require power hearing aids. This requires the use of hearing aid receivers with low impedance. Such receivers also tend to demand a high current drain that may not be sustainable by a typical hearing aid cell (or battery). Many existing power hearing aids drain at between 2.5 and 4 mA for typical gain settings and input levels. In some environments in which the input levels are high, current drain increases dramatically. This is seen in Figure 10, which shows rapid current drain above an input of 75 dB SPL when the hearing aid was adjusted for a severe-to-profound degree of hearing loss (85 to 100 dB HL). On the other hand, the drain was roughly constant at around 0.9 mA when the aid was set for a moderate degree of hearing loss (45 to 60 dB HL). This not only means a shorter battery life in the severe loss case but also the possibility of intermittent performance at high-input levels. Subjectively, the wearers may report that the hearing aid "fades in and out" or that the hearing aid is "dead." This occurs because many batteries cannot sustain the high current drain for more than a short time. Not only will such occurrences be distressing to the wearers, they could also pose safety issues and cause a loss of information.

There are two possible solutions to ensure consistent performance from hearing aids regarding the current drain issue. One is the use of "super-power" (or high-performance) batteries made by some battery manufacturers. These batteries can provide a higher current level for a longer period of time in comparison with conventional batteries. Another solution is to have the hearing aids continuously monitor current consumption to guard against excessive current drain that causes "blackouts." Some digital

![Figure 10](image-url)
high-power hearing aids have incorporated digital algorithms to serve this purpose (Kuk and Ludvigsen, 2000).

**System (or Group) Delay**

Although not all digital hearing aids are the same, many have greater flexibility and offer a better chance of ensuring consistent audibility (and performance) for their wearers. On the other hand, the very mechanism of DSP could result in artifacts. One is the group delay from the processor. In a digital hearing aid, the acoustic signal is first digitized into a series of “1” and “0” in blocks, which are then processed by the central processing unit. Depending on the size of the block and how much processing is involved, delays could be introduced in the processed signal. This means that compared with the unprocessed signal, the digitally processed signal may reach the wearer's eardrum at a later time. This does not occur in analog hearing aids.

If wearers simply hear the processed signal alone, delays as great as 15 to 20 msec may not cause any asynchrony between the visual message and the auditory message. On the other hand, for wearers with a mild hearing loss for which a vent is used in the earmold or in the hearing aid, the vent forms a direct sound path between the environment and the wearers' eardrum. Consequently, the sound that the wearer hears is a mixture of the direct sound via the vent and the processed sound from the hearing aid. If the processed sound is delayed by more than 5 to 15 msec, wearers may notice an “echoic” sensation (Stone and Moore, 1999; Agnew and Thornton, 2000). Furthermore, any difference in the arrival time of the two sound sources at the eardrum would result in cancellation of some parts of the waveform and reinforcement of the other parts. This affects sound quality and intelligibility of the input signal. When selecting digital hearing aids, it is important to ensure that their group delay is smaller than 10 msec.

**Input and Output Saturation Distortions**

Hearing aids are miniature amplifiers connected to miniature loudspeakers. When the input to the amplifiers is too high, significant saturation distortion could occur at the input stage. A distorted input will, even without additional distortion from the amplifier or loudspeakers, result in a distorted output. Poor sound quality, reduced intelligibility, and loss of audibility of the lower input sounds occur (Agnew, 1996). Fortunately, many of the current programmable and digital hearing aids include an input compressor (compression limiting) to minimize saturation distortion. This, however, could introduce temporal distortion of the high input. Other approaches, such as roving input dynamic range and extended input dynamic range, are means to prevent input distortion without introducing temporal distortion to the input signals.

At the same time, excessive output from the amplifier would result in saturation (and other) distortion at the receiver stage. Again, this would sacrifice the sound quality and may reduce intelligibility. Many current hearing aids (analog and digital) have incorporated a compression-limiting circuit to minimize such an occurrence.

**CONSISTENCY IS ENSURING THAT HUMAN FACTORS ARE CONSIDERED**

Although the factors discussed to this point may ensure that one selects hearing aids that provide consistent amplification to the child's hearing loss and listening needs at all times, non-hearing aid factors are also important to consider to ensure consistent hearing aid use. Proper maintenance such as the use of a desiccant to ensure proper functioning of the microphones and receivers and a long battery life could ensure consistent performance. In addition, proper attitudes toward the use of hearing aids and realistic expectations of the hearing aids are important factors to consider. Dillon (2001) provides a good summary of items that a clinician could perform to achieve consistent use of hearing aids with pediatric patients.

**CONCLUSION**

Because of the advances in technology and the efforts of the profession, more children with a hearing loss are identified earlier and aided with amplification. However, if hearing health care professionals are not careful in selecting hearing aids that can provide consistent performance for the child all of the time, the intended benefits from the use of amplification may not be fully realized. This is true with any form of amplification—analog, programmable, and digital. Thus, it is important for hearing health care professionals to ask specific questions on how the selected hearing aids would ensure consistent input for the pediatric wearer prior to the fitting. Do not select hearing aids simply because of their alleged advances in technology. Hearing


