Recent developments in circuit design have expanded the amplification options available to both children and adults. The complexity of these systems has resulted in a need for new and efficient fitting procedures and protocols. For adults with hearing loss, the fitting process often is supplemented with a variety of subjective measures such as judgments of loudness, clarity, or intelligibility. For obvious reasons, these types of measures cannot be obtained reliably from infants and young children.

A second fitting approach is to use a prescriptive algorithm that is based on each individual's audiometric thresholds. Numerous threshold-based procedures have been developed for use with linear hearing aids (Byrne and Dillon 1986; Cox 1988; McCandless and Lyregaard 1983; Seewald 1992) and, more recently, nonlinear hearing aids (Cornelisse, Seewald, and Jamieson 1995; Killion 1995; Dillion et al. 1998). The majority of these procedures, however, were developed using data from adults with hearing loss. Only the desired sensation level (DSL) procedure (versions 3.1 and 4.0) was designed to specifically account for the many differences between young children and adults (Cornelisse, Seewald, and Jamieson 1995; Seewald 1992). The underlying goal of this procedure is to provide an amplified signal that is both audible and comfortable across as broad a frequency range as possible. Using this procedure, all audiometric and electroacoustic data are transformed to an equivalent ear canal sound pressure level (SPL) to facilitate comparisons between audiometric results and hearing aid data. The known age-related changes in external ear characteristics (Bentler 1991; Kruger 1987; Kruger and Ruben 1987) and real-ear-to-coupler differences (Feigin et al. 1989; Nelson Barlow et al. 1988) are taken into consideration in the development of target gain and maximum output values. In addition, the speech spectrum used to derive target values takes into account the child’s need for auditory self-monitoring of speech (Cornelisse, Gagné, and Seewald 1991a). The underlying assumption is that speech must be audible in order to optimize the use of residual hearing. In this chapter, a variety of issues related to the definition of audibility will be discussed in relation to the hearing aid fitting process.

**Audibility of Speech**

For the past few decades, numerous investigators have attempted to relate a variety of psychoacoustic measures (e.g., critical band estimates, gap detection, temporal resolution) to speech recognition in both listeners with normal hearing and those with hearing impairments. In general, these studies have tended to support the notion that the audibility of speech plays the primary role in predicting performance. If a signal is inaudible, by definition, it cannot be detected, discriminated, recognized, or learned. While this appears to be a relatively obvious assumption, the measurement and quantification of audibility are actually quite complex.

**Long-Term Average Speech Spectrum (LTASS)**

One approach is to define audibility in terms of the long-term average speech spectrum (LTASS), but this can be measured in a variety of ways. The LTASS can be obtained by measuring the long-term rms spectrum produced by a single talker or multiple talkers, at various distances and azimuths, and with different types of speech materials. Values typically are taken over a 1- to 2-minute period and expressed as one-third-octave band levels when used for the prescription and evaluation of hearing aid fittings. Both the overall level and the shape of the LTASS are known to vary with speaker characteristics (Byrne 1977), the azimuth of the sound source in relation...
to the recording location (Cornelisse, Gagné, and Seewald 1991a, b; Stelmachowicz et al. 1993), distance (Pearsons, Bennett, and Fidell 1977), vocal effort (Pearsons, Bennett, and Fidell 1977), type of speech material (Studebaker and Sherbecoe 1993), and native language (Byrne et al. 1994). At the present time, there is no universally accepted version of the LTASS for clinical use although efforts are under way in this regard.

The most widespread method for computing audibility is the Speech Intelligibility Index (SII) (American National Standards Institute 1997), formerly known as the Articulation Index (American National Standards Institute 1969). SII calculations are based on the LTASS, knowledge of interfering noise levels, and estimates of hearing sensitivity as a function of frequency. The resultant values range from 0 to 1.0, where 0 represents complete inaudibility and 1.0 indicates that the entire 30 dB range of speech (at a given overall level) is audible. Appropriate transfer functions can be used to predict intelligibility for a variety of speech materials. This approach utilizes an “idealized” LTASS and assumes that the amplitude distribution of speech is 30 dB at all frequencies. Cox, Matesich, and Moore (1988) have shown that the short-term amplitude distribution of speech varies as a function of frequency and may deviate substantially from the nominal 30 dB value. These investigators also point out that these measures are highly dependent upon the short-term measurement interval used in the analysis of the speech samples.

When attempting to estimate the audibility of speech that has been processed by a hearing aid, the use of an “idealized” LTASS may produce misleading results. Given the highly complex nature of current hearing aids with respect to nonlinearities, attack and release times, and other advanced processing schemes, it may not be possible to predict the amplified spectrum accurately unless “real” speech is used as the input signal. Figure 1 shows the amplified speech spectrum as measured through a 2-channel wide dynamic range hearing aid. Running speech was presented at an overall level of 60 dB SPL and averaged over a 1-minute period using a 120 ms Hanning window with 50% overlap. The open circles represent right ear auditory thresholds from a child with a moderate-to-severe sensorineural hearing loss. These thresholds have been converted to one-third-octave band ear canal levels to facilitate comparison to the amplified speech spectrum. The solid line represents the long-term average spectrum, and the various broken lines depict

![Figure 1](image1.png)

**Figure 1.** Solid line shows the 1/3-octave band levels of the amplified LTASS as a function frequency. Open circles show thresholds for a child with a moderate-to-severe hearing loss. The broken lines depict the 10% to 90% short-term amplitude distribution of speech.

![Figure 2](image2.png)

**Figure 2.** Top left panel (a) shows thresholds for a child with a moderate hearing loss (circles) and the LTASS of Cox and Moore (1988). Top right panel (b) shows the real-ear aided gain added to the LTASS. Dashed lines in the lower left panel (c) show the 30 dB range of speech, and the crosshatched region shows the portion of the spectrum that is audible. In the lower right panel (d), the real-ear saturation response is added. (From Stelmachowicz 1996.) Reprinted with permission. ©1996. *Amplification for children with auditory deficits* (pp. 193-213). Nashville, Tenn.: Bill Wilkerson Center Press.
the 10% to 90% short-term amplitude distribution of speech. Specifically, 90% of the short-term levels exceeded the lowest broken line, and less than 10% fell above the upper broken line. In theory, using real speech to measure hearing aid output should provide the most valid representation of the amplified LTASS because the time-dependent system nonlinearities of the hearing aid will be operating as they would in normal use.

Unfortunately, this type of approach would be prohibitive in a typical clinical setting because of the time involved (approximately 1 minute per frequency response). While we have found that the LTASS values can be obtained reliably when averaging for periods as short as 15 sec, the short-term amplitude distributions are not accurately represented under these circumstances, and 15 sec/response would still be considered excessive for clinical purposes relative to current hearing aid test systems that can provide frequency responses in real time. Thus, alternative strategies are needed to characterize the amplified spectrum for clinical use. Figure 2 illustrates one such method. All values in this figure are expressed in SPL as measured in the ear canal. In the top left panel (a), the open circles depict pure-tone thresholds for an individual with a moderate hearing loss, and the solid line represents the LTASS as described by Cox and Moore (1988). The top right panel (b) shows the real-ear aided gain added to the LTASS to yield the amplified spectrum, shown by the dashed lines. In the lower left panel (c), the nominal +15 dB range of speech (American National Standards Institute 1997) is added to the amplified spectrum, and the crosshatched region depicts the portion of the signal that is audible. In the final panel (d), the real-ear saturation response is shown by the asterisks to provide information about the aided dynamic range of speech. An SII could be easily calculated from this information.

Variations of this approach have been automated for clinical use (Seewald et al. 1997; Stelmachowicz, Kalberer, and Lewis 1996). Hearing aid gain and maximum output may be measured either in a 2cc coupler or in the real ear using a probe microphone. While this approach is clinically feasible, recent data suggest that the type of stimuli used to measure gain may influence the results when the hearing aid is operating in nonlinear fashion (Stelmachowicz et al. 1990; Stelmachowicz et al. 1996). In the Stelmachowicz et al. (1996) study, real speech and a variety of other stimuli were used to measure hearing aid gain of both linear and nonlinear systems. When a hearing aid was functioning within its linear operating range, stimulus type had little influence on the measured gain. For inherently nonlinear systems (e.g., wide dynamic range compression) or linear hearing aids operating at high input levels, significant differences in the measured gain were found across stimuli. These differences increased as both a function of frequency and input stimulus level. Figure 3 shows the difference between the gain measured using real speech and the gain measured using either swept pure tones (left) or speech-weighted composite noise (right) as a function of input level for four different nonlinear hearing aids. Data show the gain at 3000 and 4000 Hz only, which is where the largest differences occurred. All values lie above zero, indicating that the gain for real speech exceeded that measured using the simpler stimuli. On average, the pure-tone stimuli showed slightly larger deviations than the composite noise stimulus. There appear to be large individual differences across these four hearing aids, with differences exceeding 10 dB in some cases.

![Figure 3. Difference between gain measured with real speech and either swept pure tones (left) or speech-weighted composite noise (right) for four nonlinear hearing aids. Open and filled symbols show data for 3k and 4kHz, respectively. Reprinted by permission of Lippincott Williams & Wilkins. ©1996. Ear and Hearing 17:520-527.](image-url)
results have important clinical implications. If pure tones were used to match target gain values, the actual gain achieved for real speech could be 5 to 10 dB higher. This may cause signal distortion due to a decrease in headroom and/or may pose a risk to residual hearing (Macrae 1994).

Figure 4, taken from the same study, shows the difference between speech gain and gain estimated with three different stimuli (speech-modulated noise, simulated speech,¹ and speech-weighted warble tones) as a function of frequency. The same four nonlinear hearing aids were used, and each panel shows results for a different input level. In general, these stimuli tend to produce results that approximate the gain for real speech more closely than either pure tones or composite noise. At the highest input level, however, differences in the order of 10 dB still occur. Test time for these stimuli is considerably longer than for the simpler stimuli.

From a clinical perspective, these results suggest that, for nonlinear processing conditions, measured gain may differ substantially when different instrumentation and stimuli are used. At this point, it is not clear which type of stimuli provides the most valid representation of speech gain, although it does appear that signals shaped to simulate the LTASS may be more appropriate than signals that are constant as a function of frequency. Clearly, more work is needed to determine the most efficient and valid test stimulus for clinical use.

Functional Gain Measures

To this point, the discussion has been limited to estimates of audibility obtained from either 2cc coupler or real-ear measures of gain. It is important to consider whether functional gain measures can be used to estimate the audibility of speech. For purposes of this discussion, functional gain will be defined as the difference between unaided and aided sound field thresholds as a function of frequency. The comments, however, also will apply to aided sound field thresholds in isolation. Prior to the clinical use of probe microphone measures, functional gain or aided sound field thresholds often were used to characterize hearing aid performance. The many limitations of this approach have been described extensively by others (Macrae 1982a, b; Stelmachowicz and Lewis 1988), and many of those details will not be reiterated here. As discussed earlier, hearing aid gain estimates will be highly dependent upon stimulus input level whenever a hearing aid is operating in a nonlinear fashion. Nonlinearities can occur intentionally (i.e., for WDRC or compression limiting systems) or unintentionally (i.e., for linear systems in saturation). Furthermore, the input levels at which nonlinearities begin to occur are likely to vary as a function of frequency. Figure 5 illustrates how this issue would impact functional gain measures. In this figure, taken from a paper by Lewis (1997), 2cc coupler gain at 2000 Hz is shown as a function of input level for three different types of hearing aids. Hearing aid A is a WDRC circuit with a compression threshold of 40 dB SPL, hearing aid B is a WDRC circuit with a compression threshold of 60 dB SPL, and hearing aid C represents either a compression limiter or a peak clipper. Assume that a child with a moderate hearing loss is being evaluated with these three types of hearing aids. Assume further that the aided sound field thresholds for each of these hearing aids will fall in the 20 to 30 dB HL (~30 to 40 dB SPL) range. Figure 5 shows that, for input signals < 40 dB SPL, the gain of the three systems is identical. For average conversational speech, which occurs at approximately 60 dB SPL, hearing aid A will provide 10 dB less gain than the other two circuits. For speech inputs of 70 dB SPL (raised voice), there are even larger differences across the three

¹The simulated speech signal was composed of a randomly selected series of 400 tones (250-8000 Hz) that were frequency-modulated at 36 Hz (+5%) to create a more speechlike stimulus. The relative amplitude and duration of the tones were designed to approximate the temporal characteristics of continuous discourse (Cole 1996).
circuit. None of these processing differences will be reflected in functional gain measures because the input levels are likely to be below the compression threshold of these instruments.

From a clinical perspective, this means that functional gain measures cannot be used to assess hearing aid performance for any device that operates in a nonlinear manner. This would include not only WDRC circuits and FM systems, but also many linear devices because these circuits often may operate nonlinearly for conversational level speech inputs. It is inappropriate to assume that functional gain measures can in any way predict hearing aid performance for typical speech input signals. To accurately estimate audibility of speech, it is essential that the input level to the hearing aid reflect the expected input level of speech. It is also important to measure hearing aid gain at multiple input levels in order to ensure that speech will be audible and comfortable over a wide range of levels. These goals can be accomplished best by using either 2cc coupler or probe microphone measures.

Short-Term Components of Speech

While audibility of the LTASS is essential if speech is to be recognized, in actual operation, most hearing aids will respond to the short-term components of speech. This is particularly true for fast-acting or syllabic compression systems. Figure 6 illustrates this point for the time waveform of the utterance /ap/ (Stelmachowicz et al. 1995). Here, the vowel is followed by a gap and then the burst that denotes the plosive. As expected, the vowel energy is considerably greater than the burst energy. If this signal were recorded through a hearing aid circuit with fast time constants, the vowel might engage compression, but the gap would be long enough for recovery from compression to occur. In such instances, the consonant-to-vowel ratio may be altered, and audibility (and possibly perception) of the /p/ might be improved. If the utterance were /ish/, however, no gap exists between the offset of voicing and the presence of the frication noise. In this case, no changes in consonant-to-vowel ratio would occur, even for fast-acting circuits. Thus, it is likely that the dynamic properties of any hearing aid circuit will interact with the dynamic properties of running speech. As such, measures of audibility based upon the LTASS may not accurately reflect the many processing differences that exist across hearing aids. Figure 7, also taken from Stelmachowicz et al. (1995), illustrates this point. In this
figure, audibility of the syllable /ap/ is shown for two different hearing aids at three input levels. The frequency response of the two hearing aids (linear peak clipper and a WDRC hearing aid) was adjusted to be equivalent with a 60 dB SPL input signal. The open circles in each panel show the auditory thresholds for a child with a moderate hearing loss. The dashed lines show the amplified spectrum of a 100 ms segment taken from the middle of the steady-state vowel (/a/), and the solid line shows the spectrum of a 10 ms segment at the onset of the burst (/p/). The percent correct score for this phoneme on a closed set nonsense syllable task is shown for each condition in the lower right corner of each panel. Because the frequency response of the two hearing aids was equated as closely as possible at 60 dB SPL, the vowel spectra at 65 dB SPL are very similar for the two hearing aids. In contrast, audibility of the /p/ for the nonlinear circuit is greater than for the linear circuit at both 50 and 65 dB SPL. This suggests that the nonlinear circuit was operating as a syllabic compressor, providing greater gain for the low-level /p/ than for the higher level /a/. The improved audibility for the nonlinear circuit at 65 dB SPL is reflected in the performance scores.

At 80 dB SPL, where the burst is clearly audible for both systems, performance is 100%. At 50 dB SPL, large differences in performance are observed, even though the audibility differences appear to be minimal. This observation raises an important issue regarding the relevant cues for perception. In this example, only the steady-state portion of the vowel and the onset spectra of the burst were considered. It is well known that there are multiple cues to speech perception, and the relative importance of these cues may vary across speech stimuli, talkers, and listening conditions. In this example, it is possible that acoustic cues in the transition from vowel to
consonant may have contributed to the performance differences at the lowest input level.

From a clinical perspective, it may be useful to combine information derived from the amplified LTASS with knowledge about the spectral envelope of specific speech sounds as described by Boothroyd, Erickson, and Medwetsky (1994). Figure 8 illustrates how this might be accomplished. In this figure, the open circles show thresholds for a child with a moderate hearing loss. The solid line shows the amplified LTASS, and the dotted lines show the 30 dB range of speech. The horizontal bars show the peak energy and frequency spread for three different final consonants (presented at 60 dB SPL) after processing by a linear hearing aid. While the amplified LTASS shows good audibility in the 2000 to 4000 Hz region, the final /t/ in the word *boot* is not audible. Providing this type of information for specific classes of speech sounds may be useful to audiologists as they fine-tune a hearing aid. That is, changes in hearing aid gain, compression threshold, or bandwidth might be used to improve the audibility of specific speech sounds.

The above discussion illustrates that the audibility of speech after processing by a hearing aid may not be easily predicted from traditional electroacoustic measures of hearing aid performance. For individuals with normal hearing or mild-to-moderate hearing losses, the ability to use multiple cues may compensate for a reduction in audibility in certain frequency regions. For infants and children with prelingual hearing loss, however, the loss of audibility may have a more significant effect on perception. Additional studies are needed to determine the relation between the available acoustic information and perception.

How Audible Is Audible Enough?

Studies with adults with hearing loss have shown that, as long as the majority of the speech spectrum is audible, changes in the relative frequency response do not result in large changes in performance (Horwitz, Turner, and Fabry 1991; Sullivan et al. 1988). Transfer functions, relating audibility to performance, for listeners with normal hearing are available for a wide range of speech materials. In recent years, a number of studies have addressed the issue of audibility in infants and young children. Compared to adults, children have been shown to demonstrate higher behavioral detection thresholds (Olsho et al. 1988) and to require higher signal-to-noise ratios (Allen et al. 1989; Schneider et al. 1989). Child-adult differences also have been reported for phoneme and word recognition tasks (Nozza, Rossman, and Bond 1991; Nozza et al. 1990; Hnath-Chisolm, Laipply, and Boothroyd 1998).

Recently, we investigated the relation between audibility and stimulus context for children in the 5- to 7-year age range and for adults. In this study, children were asked to repeat 120 four-word sentences that were either high predictability (HP), *"Stay off the hill,"* or low predictability (LP), *"Cups give fat ducks."* Test materials were adapted from those developed by Boothroyd and Nittouer (1988). Stimuli were presented in quiet at five presentation levels in order to construct performance-intensity functions. Within each age group, a best fit relating recognition to the Audibility Index (AI) was obtained. Figure 9 shows the AI value corresponding to 80% correct word recognition as a function of age for both the HP and LP sentences. Results revealed clear age-related changes suggesting that young children require a higher AI than adults for equivalent performance. In addition, slightly higher AI values were needed for the LP sentences where semantic context was limited. These age-related differences were most apparent at low sensation levels. These findings have implications for hearing aid selection and fitting protocols with young children. Since performance differences between children and adults appear to be largest for low levels of audibility, it would...
seem that wide dynamic range compression might work well in children by providing greater amplification of low-level input signals. Additional studies are needed with a broader range of children with normal hearing or hearing impairment to determine the magnitude of child-adult differences at lower age levels and the age at which adult like performance is achieved.

Summary and Recommendations for Clinical Practice and Future Research

So, are our current clinical procedures for fitting hearing aids to infants and young children correct? The answer is: probably not. As always seems to happen when knowledge is gained in a specific area, more questions arise than answers. Good clinical practice in any area involves a constantly changing set of protocols and guidelines based upon the best available knowledge from well-controlled studies. Incremental gains in knowledge often reveal “what not to do” rather than the “best approach” to a specific problem. Based upon the studies reviewed in this chapter, the following clinical recommendations are proposed:

1. It is probably incorrect to assume that hearing aid fitting algorithms based on adult data will be optimal for young children who are in the process of developing speech and language.

2. Since audibility is known to be essential for speech and language development, an audibility-based algorithm should be used to ensure that the amplified LTASS is audible over as broad a frequency range as is possible.

3. Given current hearing aid technology and what we know about the listening environment of children, audibility of amplified speech should be assessed at multiple input levels.

4. WDRC may be an appropriate way to compensate for children’s reduced performance at low sensation levels without posing a risk to residual hearing.

5. Audibility of the LTASS does not ensure audibility of specific speech sounds.

6. Functional gain measures cannot provide a valid representation of performance for most hearing aid fittings.

In order to improve our management of the pediatric population, additional research studies are needed in the following areas:

1. Studies are needed to gain a better understanding of developmental trends in the area of audition and how these might relate to the process of fitting hearing aids. This work should include studies of audibility, speech perception in noise and reverberation, acclimatization, and lexical learning.

2. Studies are needed to determine the optimal signals for evaluating the electroacoustic characteristics of hearing aids. These signals should provide gain measures that closely approximate the gain for running speech in as short a time period as possible. Techniques should be evaluated with a wide range of hearing aid processing schemes in order to assure the robustness of the evaluation process.

3. Studies are needed to characterize the audibility of specific speech sounds and relate these findings to perception in young children.

References


