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Effect of Preferred Volume Setting on Speech Audibility in Different Hearing Aid Circuits

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Abstract

This study compared preferred volume setting for linear peak clipping, compression limiting, and wide dynamic range compression (WDRC) amplification and quantified speech audibility at the preferred volume setting for each amplification type. Ten listeners with mild-to-moderate hearing loss were fitted monaurally with a behind-the-ear hearing aid programmed sequentially with WDRC, compression limiting, and linear peak clipping amplification. Speech was presented in quiet and in noise at a range of input levels. In each condition, the listener adjusted the volume for maximum clarity. Signal levels were measured using a probe microphone system. There was no significant difference in speech audibility between amplification strategies for any speech level regardless of the presence or absence of background noise. These results imply that the improved audibility advantage of WDRC over linear amplification demonstrated in controlled environments may not be maintained in situations where the hearing aid wearer controls the volume.

Key Words: Amplification, audibility, clarity, preferred volume

Abbreviations: AAI = Aided Audibility Index, AI = Articulation Index, ANOVA = analysis of variance, NAL-R = National Acoustic Laboratories-Revised, SIR = Speech Intelligibility Rating, WDRC = wide dynamic range compression

The increased use of wide dynamic range compression (WDRC) amplification has generated considerable discussion of its benefits relative to linear amplification. The theoretical advantage of WDRC is improved audibility of low-level speech cues without discomfort from high-level sounds. However, the results of studies with wearable aids have been mixed: some studies demonstrate superior speech recognition performance with WDRC aids, whereas others show limited or no benefit from this strategy (see Hickson, 1994, and Dillon, 1996, for reviews).

One possible influence on these results is that the theoretical benefits of improved audibility may not be maintained in situations in which the listener controls the hearing aid volume setting. In part, this argument is the basis for the exclusion of volume controls in many WDRC hearing aids. However, about half of hearing aid wearers fit with WDRC aids that do not include a manual volume adjustment would prefer to have a volume control (Knebel and Bentler, 1998; Valente et al, 1998; Kochkin, 2000). It is important to understand the effects of such volume adjustments, which can alter the relative output levels of speech amplified by different hearing aid circuits.

Several recent studies support the idea that variations in volume setting can significantly affect the advantage of improved audibility offered by WDRC amplification. In fact, the benefits of WDRC have been demonstrated most clearly in controlled studies in which the experimenter sets the amount of gain and thus the audibility of the speech signal. In this situation, WDRC can improve speech audibility, with corresponding improvements in speech recognition, over linear amplification. For example, when test conditions are designed to provide greater audibility for WDRC-amplified than for linearly amplified speech, speech recognition is higher with the WDRC hearing aid (Moore and Glasberg, 1986; Souza and Turner, 1998). However, when speech audibility is equated between

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hearing aids, listeners show similar performance for linear versus WDRC amplification (Souza and Turner, 1996, 1998). The recent National Institute on Deafness and Other Communication Disorders/Veterans Affairs clinical trial found minimal differences in speech recognition between WDRC and linear (compression limiting or peak clipping) hearing aids when hearing aid volume was adjusted to National Acoustic Laboratories-Revised (NAL-R) targets (Larson et al, 2000). Presumably, NAL-R targets were similar across amplification types for moderate input levels, resulting in similar speech audibility and no net advantage for the WDRC hearing aid in that situation.

In comparisons of compression and linear amplification where the listener adjusted the volume, the results were mixed: some investigators found superior performance with the WDRC aid (e.g., Laurence et al, 1983), whereas others found no difference between circuits (Dreschler et al, 1984; Tyler and Kuk, 1989). Because speech output level at the preferred volume was not specifically measured in these studies, it is difficult to distinguish between the effects of volume adjustment on speech audibility and other procedural factors (e.g., frequency response, compression parameters). However, the studies described above do suggest that when the listener has control over the volume setting, a listener who selects a higher volume setting for WDRC processing than for linear processing may show better performance with the WDRC aid, whereas a listener who selects similar output levels for both aids may show no difference in recognition.

There is considerable information available about how listeners select gain for linear hearing aids (Cox, 1984; Byrne and Cotton, 1987; Kuk, 1990; Leijon et al, 1990; Cox and Alexander, 1991; Kuk and Lau, 1996; Humes et al, 2000; Scollie et al, 2000). Preferred volume setting depends on the listening environment, including talker level, signal-to-noise ratio, amount of reverberation, and degree of distortion (Walden et al, 1977; Cox and Alexander, 1994). For example, linear hearing aid wearers select higher gain in quiet than in noisy or reverberant environments (Cox and Alexander, 1991). Overall, patients select lower volume levels for everyday listening than those prescribed by clinicians (Leijon et al, 1984, 1990; Cox and Alexander, 1991; Humes et al, 2000).

We know less about how a listener adjusts gain for WDRC hearing aids. In a recent study, Neuman et al (1995) compared preferred listening levels for slow-acting compression versus linear laboratory-based amplification systems. Input-output functions were equated relative to a 70-dB input signal presented at the most comfortable level, providing successively higher gain for low-intensity speech as compression ratio increased. At low compression ratios, preferred listening levels were similar for linear and slow-acting compression amplification. At higher compression ratios, the listeners selected higher listening levels for the linear amplification. Because this study used long time constants and a higher kneepoint than typical for WDRC systems, it is not possible to apply these results directly to WDRC hearing aids. Also, because a laboratory-based system was used, many of the effects present in wearable aids (e.g., earmold and venting effects, distortion owing to output limiting) were not incorporated. Such effects are important to consider because they can influence volume setting (Kuk, 1990).

To explore these issues, the goals of this study were to (1) compare the subject's preferred volume setting for linear peak clipping, compression limiting, and WDRC amplification and (2) to quantify speech audibility at the patient's preferred volume setting for each type of amplification.

METHOD

Participants

Participants were 10 listeners (7 female, 3 male), aged 23 to 70 years (mean = 52 years), with bilateral sensorineural hearing loss. All were native speakers of English. Six of the listeners were current binaural hearing aid wearers; of these, 3 used compression-limiting hearing aids, 2 WDRC hearing aids, and 1 a linear peak clipping aid. Three had never worn amplification and 1 was a prior hearing aid user, but the aid was not available for testing. Because listener-selected gain appeared to be stable over time for both the new and experienced hearing aid users (Leijon et al, 1990; Cox and Alexander, 1992; Bentler et al, 1993; Humes et al, 2000), the participants were considered a single test group without regard to prior amplification use. Frequency-specific loudness discomfort levels were measured for warble tones presented at 0.5, 1, 2, and 4 kHz via an ER-3 insert earphone. As recommended by Hawkins et al (1987), loudness discomfort level was defined as the lowest intensity that evoked a response of "uncomfortably loud" in two of three ascending trials, minus

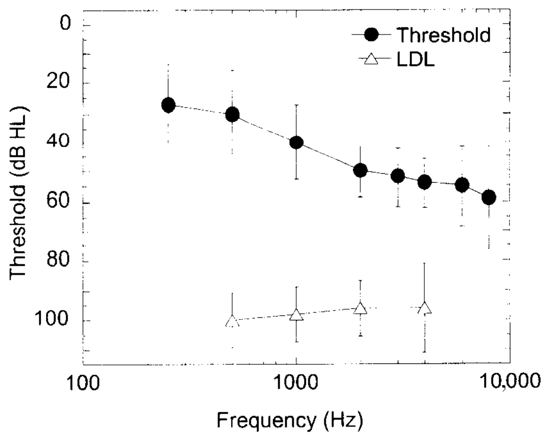

Figure 1 Mean hearing thresholds and loudness discomfort levels (LDL). Error bars represent ± 1 SD about the mean.

1 dB. The mean pure-tone thresholds and loudness discomfort levels are shown in Figure 1.

Hearing Aid

During the test sessions, each listener was fitted monaurally with a programmable, multi-memory, behind-the-ear hearing aid with omnidirectional microphone and remote control. In cases of asymmetric loss, the better ear was selected. In cases of symmetric loss, one ear was selected at random. The nontest ear was occluded with a foam earplug. This method was purposely used rather than introduction of masking noise to the nontest ear because the presence of noise can influence the selection of the preferred volume setting (Cox and Alexander, 1991). The hearing aid was programmed sequentially (in random order) with WDRC, compression limiting, and linear peak clipping amplification. The hearing aid frequency response, output sound pressure level with 90 dB SPL input, and WDRC ratio were individually set for each listener according to the Desired Sensation Level (Cornelisse et al, 1995) prescriptive procedure and programmed using the manufacturer's fitting software. For WDRC amplification, the hearing aid operated as an input compressor with compression threshold fixed at 50 dB SPL and compression ratios ranging from 1:1 to 2.7:1. For compression-limiting amplification, the hearing aid operated as an output compressor with compression thresholds between 65 and 85 dB SPL and compression ratios between 8:1 and 20:1, depending on the volume setting. Representative output and gain functions for the three hearing aid circuits are shown in Figure 2.

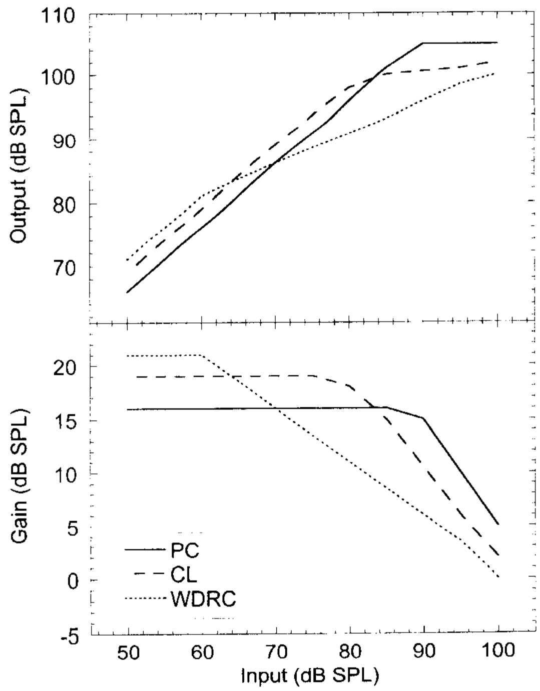

Figure 2 Comparison of representative output (*top panel*) and gain (*lower panel*) functions for the three hearing aid circuits. All values are 2-cc coupler levels for a 2-kHz input signal with the hearing aid programmed for the mean participant audiogram. PC = peak clipping; CL = compression limiting; WDRC = wide dynamic range compression.

The hearing aid was coupled to a lucite skeleton earmold with select-a-vent. Target match was verified using probe microphone measurements, and the frequency-gain response and/or vent size was adjusted if necessary to obtain the closest possible match to real-ear insertion gain target (Fig. 3).

Materials

The test material was a compact disc recording of the Speech Intelligibility Rating (SIR) test, developed by Cox and McDaniel (1989). The SIR test consists of 20 passages on everyday topics, adjusted for equal intelligibility. The SIR test also contains a multitalker babble with spectrum matched to the speech spectrum. The speech and noise outputs from the compact disc player (Tascam CD301) were routed to separate attenuators (TDT PA4), mixed (TDT SM3), amplified (Crown D75), and played through a speaker (Optimus XTS36) for presentation to the listeners. The speaker was placed at head level, 1 meter in front of the listener at 0 degrees azimuth in a double-walled sound booth.

Speech was presented at three input levels: 50, 65, and 80 dB SPL, representing low, moderate, and high-level speech, respectively. Sound levels were root mean square levels specified at the position corresponding to the center of the listener's head with the subject removed from the sound field and were maintained through daily calibrations. Each speech level was presented in quiet and in noise at a +7 signal-to-noise ratio for a total of six conditions.

Procedure

For each subject, the order of the three amplification strategies was counterbalanced. The order of the six test conditions was randomized within each type of amplification. In each condition, the listener was instructed to adjust hearing aid volume, using the remote control, for maximum speech clarity. Specifically, the listener was instructed to set the hearing aid to a level that would allow them to follow the conversation without strain. Clarity was chosen as the criterion rather than overall preference because hearing aid wearers describe this to be the most important property of amplified speech (Hagerman and Gabrielsson, 1985). The listener was allowed to adjust the volume freely until the criterion was reached. Typically, the desired volume setting was reached within one passage; however, the listener was allowed to listen to as many passages as desired. The remote control volume adjustment offered a range of 26 dB with a 1.3-dB step size. If the listener reached the top or bottom limit of the range before achieving the desired volume setting in any condition, the overall gain was reprogrammed to allow the listener to reach the desired level.

When the listener indicated that the preferred volume setting had been reached, output of the hearing aid at the tympanic membrane was measured using a probe microphone system (Etymotic Research 7C) routed to a spectrum analyzer (HP35665A). Probe microphone rather than coupler measurements were used because they provide a more accurate measure of sound pressure level (Barlow et al, 1988). The test signal for probe microphone measurements was the speech-shaped calibration noise provided on the compact disc recording of the SIR test. This noise is shaped to have the same long-term average spectrum as the test stimulus. Without altering the test environment (i.e., speaker position and azimuth or hearing aid volume setting), overall root mean square levels of the

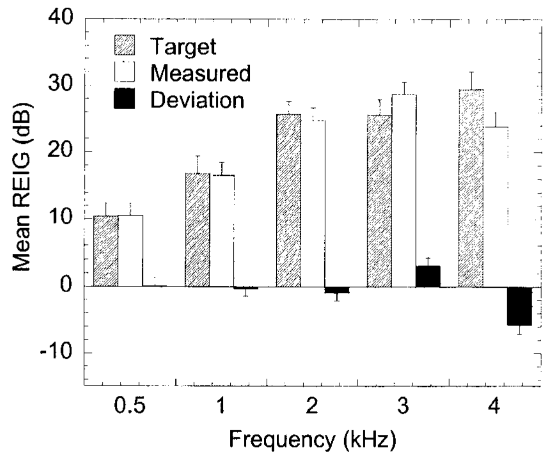

Figure 3 Mean target (*shaded bars*) and measured (*open bars*) real-ear insertion gain (REIG). Dark solid bars show the mean deviation from target at each frequency. Error bars represent one standard error of the mean.

signal were measured for each volume trial. One-third octave band levels were also obtained for use in the speech audibility calculations described below. Two trials were conducted in each test condition and averaged for analysis. Test–retest variation was approximately 1 dB across trials. The hearing aid volume was reset to minimum after each trial.

RESULTS

Preferred Volume Setting

The measured output at the listeners' preferred volume setting is shown in Figure 4. The top panel shows the results in quiet and the lower panel shows the results in noise. The results were analyzed using a three-way, repeated-measures analysis of variance (ANOVA). Within-subject factors were amplification type (linear peak clipping, compression limiting, WDRC), speech level (50, 65, 80 dB SPL), and signal environment (presence or absence of background noise).

The three-way interaction was nonsignificant ($F = 1.65$, $df = 4, 36$, $p = .183$). There was no interaction between the speech input level and the signal environment (quiet or noise) ($F = 1.54$, $df = 2, 18$, $p = .241$) or between input level and amplification type ($F = 1.23$, $df = 4, 36$, $p = .314$). The main effect of input level was nonsignificant ($F = 1.05$, $df = 2, 18$, $p = .370$). In other words, patients adjusted gain from approximately 4 dB for loud speech to 35 dB for soft speech to provide a constant output across a

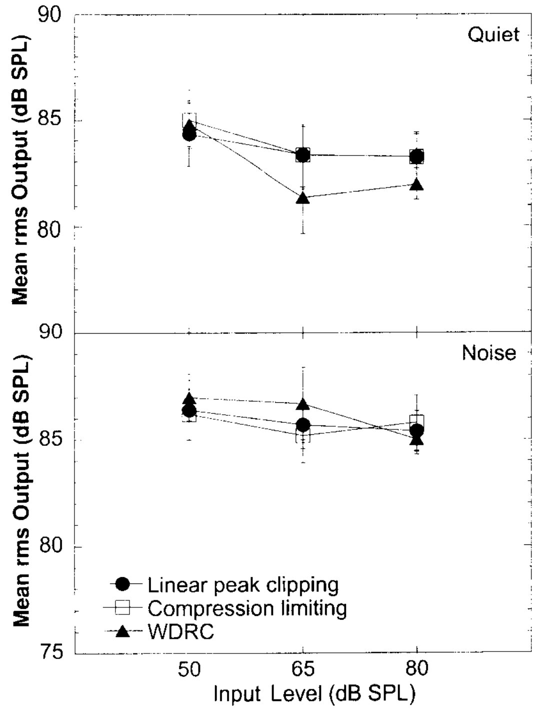

Figure 4 Mean root mean square (rms) levels measured at the tympanic membrane with the aid set to the preferred volume for each amplification type. The top panel shows the results in quiet and the lower panel shows the results in noise. Error bars represent ± one standard error above the mean. WDRC = wide dynamic range compression.

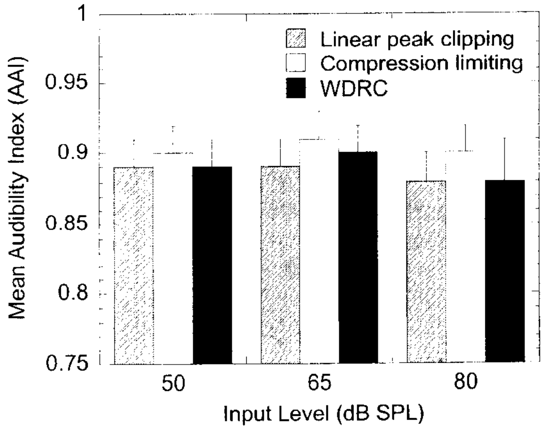

Figure 5 Comparison of speech audibility, calculated at the preferred volume setting, for each type of amplification and speech input level. Error bars represent one standard error above the mean. AAI = Aided Audibility Index.

range of input levels, regardless of the type of amplification or the presence of background noise.

There was a significant interaction between the type of amplification and the signal environment ($F = 8.51$, $df = 2, 18$, $p = .003$). Simple main effects were analyzed further by completing separate one-way, repeated-measures ANOVAs comparing the type of amplification within each signal environment (quiet or noise). The results indicated that listeners chose similar output levels for each amplification type in quiet ($F = 3.08$, $df = 2, 18$, $p = .071$) and in noise ($F = .54$, $df = 2, 18$, $p = .594$). For each type of amplification, listeners selected greater gain (approximately 3 dB more) in noise than in quiet ($p < .005$).

Speech Audibility

A second goal in this study was to examine speech audibility at the listener's preferred volume setting. Because differences in amplification processing can affect the extent to which low-intensity cues are amplified relative to high-intensity cues, the selection of similar long-term speech output levels across amplification types does not guarantee equivalent audibility of the short-term intensity range. Put another way, the preferential amplification of lower-intensity signals by a WDRC circuit likely increases the overall audibility of the signal over that of a linear circuit.

The ⅓-octave band levels measured at the tympanic membrane in the quiet condition were used to calculate Aided Audibility Index (AAI) values (Stelmachowicz et al, 1994). The AAI is an importance-weighted measure of audibility that indicates the proportion of the speech amplitude range available to the listener. Calculation of the AAI is similar to the Articulation Index (AI; French and Steinberg, 1947; Fletcher and Galt, 1950) and its counterpart, the Speech Intelligibility Index (ANSI, 1995). In contrast to those measures, which assume a 30-dB range for linearly amplified speech without distortion (i.e., peak clipping), the AAI calculation also accounts for the effects of output limiting and for the reduction of the speech amplitude range owing to compression. As for the AI, an AAI of 0.0 indicates that the entire range of speech levels is below the listener's threshold. An AAI of 1.0 indicates that the entire range of speech levels is audible to the listener. Several recent articles provide a detailed description of the AAI calculation (Stelmachowicz et al, 1998; Souza and Turner, 1999; Souza and Bishop, 2000).

Mean AAI values at the listener's preferred volume settings in quiet are shown in Figure 5. An ANOVA was used to compare speech audibility as a function of amplification type and speech input level. There was no interaction between amplification type and input level ($F = .44$, $df = 4, 36$, $p = .782$). Speech audibility did not vary as a function of speech input level ($F = .34$, $df = 2, 18$, $p = .718$) or type of amplification ($F = 2.81$, $df = 2, 18$, $p = .087$).

DISCUSSION

When asked to set their hearing aid volume to maximize speech clarity, listeners with mild-to-moderate loss set the hearing aid volume to provide a listening level (measured at the tympanic membrane) of approximately 83 dB SPL, regardless of input level. This pattern of increasing gain with decreasing input level agrees with previous research by Cox and Alexander (1994) for linear amplification and Neuman et al (1995) for slow-acting compression hearing aids. In the present study, this produced a small, statistically equivalent range of output levels for low, moderate, and high input levels, whereas previous research indicated slightly greater output for higher input levels.

For speech in noise, listeners increased gain by approximately 3 dB over speech in quiet to maintain clarity. This finding conflicts with some previous work that suggested a lower volume setting for speech in noise versus speech in quiet (Cox and Alexander, 1991). Although Neuman et al (1995) also demonstrated a general trend toward decreasing gain as the noise level increases, the results showed considerable variability. For example, measures of speech presented in a background of cafeteria noise, the background noise most similar to our study, showed a U-shaped pattern, with the lowest gain for intermediate signal-to-noise ratios and the higher gain at higher and lower signal-to-noise ratios.

In the present study, a criterion of maximum clarity rather than comfort likely caused the subjects to increase the volume in the presence of background noise. Anecdotally, some listeners noted that they would have adjusted the volume differently given different instructions. Clarity was specifically chosen as the criterion because hearing aid listeners rate this as the most important characteristic of their fittings. In everyday listening situations, the criterion for adjusting volume may vary with the level of background noise. For example, Barker and Dillon (1999) suggested that comfort may be relatively more important than speech intelligibility in high levels of noise. Kuk et al (1994) also emphasized clarity as a criterion for volume adjustment but found that subjects selected less gain as background noise levels worsened past a signal-to-noise ratio of +5 dB. However, in that study, listeners adjusted low-frequency gain independently. Manipulation of the lower frequencies may have effectively reduced the influence of background noise, resulting in an outcome similar to increasing the overall level of the signal.

There was no significant difference in speech audibility (i.e., AAI) between hearing aid processing strategies for any speech level regardless of the presence or absence of background noise. Although speech recognition was not directly measured in this study, equivalent AAIs across circuit types predict equivalent recognition scores across circuit types (Souza and Turner, 1999). Because lower compression thresholds are generally associated with higher gain for low-intensity sounds, they are presumed to improve recognition of low-intensity consonants (Dreschler, 1988; Kuk and Ludvigsen, 1999) and thus increase recognition scores over linear amplification. Instead, our results imply that the advantage of improved audibility over linear amplification demonstrated with WDRC aids in a restrictive laboratory environment may not be maintained in more realistic situations, when the hearing aid user has control over the precise gain.

Perhaps more importantly, even for WDRC amplification that theoretically reduces the need for a volume control, listeners required a large adjustment in volume setting as the presentation level changed to maintain maximum clarity. These results support those noted by Barker and Dillon (1999), for whom the majority of subjects wearing low compression threshold aids reported using their volume controls. This is an important point, particularly given the lack of manual volume controls in many WDRC hearing aids and the stated preference for manual volume controls in previous work (Knebel and Bentler, 1998; Valente et al, 1998; Kochkin, 2000). It is unclear to what extent this preference is influenced by previous experience with linear versus WDRC amplification as there are few data available on acclimatization to circuit type. Future work should be expanded to consider these issues.

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REFERENCES

American National Standards Institute. (1995). Methods for the calculation of the speech intelligibility index. (ANSI S3.5-1995). New York: ANSI.

Barker C, Dillon H. (1999). Client preferences for compression threshold in single-channel wide dynamic range compression hearing aids. *Ear Hear* 20:127–139.

Barlow N, Auslander M, Rines D, Stelmachowicz P. (1988). Probe-tube microphone measurements in hearing-impaired children and adults. *Ear Hear* 9:243–247.

Bentler R, Niebuhr D, Getta J, Anderson C. (1993). Longitudinal study of hearing aid effectiveness. I: objective measures. *J Speech Hear Res* 36:808–819.

Byrne D, Cotton S. (1987). Preferred listening levels of sensorineurally hearing-impaired listeners. *Aust J Audiol* 9:7–14.

Cornelisse LE, Seewald RC, Jamieson DG. (1995). The input/output formula: a theoretical approach to the fitting of personal amplification devices. *J Acoust Soc Am* 97:1854–1864.

Cox RM. (1984). Relationship between aided preferred listening level and long-term listening range. *Ear Hear* 5:72–76.

Cox RM, Alexander GC. (1991). Preferred hearing aid gain in everyday envinronments. *Ear Hear* 12:123–126.

Cox RM, Alexander GC. (1992). Maturation of hearing aid benefit: objective and subjective measurements. *Ear Hear* 13:131–141.

Cox RM, Alexander GC. (1994). Prediction of hearing aid benefit: the role of preferred listening levels. *Ear Hear* 15:22–29.

Cox RM, McDaniel R. (1989). Development of the Speech Intelligibility Rating (SIR) test for hearing aid comparisons. *J Speech Hear Res* 32:347–352.

Dillon H. (1996). Compression? Yes, but for low or high frequencies, for low or high intensities, and with what response times? *Ear Hear* 17:287–307.

Dreschler WA. (1988). The effect of specific compression settings on phoneme identification in hearing-impaired subjects. *Scand Audiol* 17:35–43.

Dreschler WA, Eberhardt D, Melk PW. (1984). The use of single-channel compression for improvement of speech intelligibility. *Scand Audiol* 13:231–236.

Fletcher H, Galt RH. (1950). The perception of speech and its relation to telephony. *J Acoust Soc Am* 22:89–151.

French NR, Steinberg JC. (1947). Factors governing the intelligibility of speech sounds. *J Acoust Soc Am* 19:90–119.

Hagerman B, Gabrielsson A. (1985). Questionnaires on desirable properties of hearing aids. *Scand Audiol* 14:109–111.

Hawkins DB, Walden BE, Montgomery A, Prosek RA. (1987). Description and validation of an LDL procedure designed to select SSPL90. *Ear Hear* 8:162–169.

Hickson LMH. (1994). Compression amplification in hearing aids. *Am J Audiol* 3:51–65.

Humes LE, Barlow NN, Garner CB, Wilson DL. (2000). Prescribed clinician-fit versus as-worn coupler gain in a group of elderly hearing-aid wearers. *J Speech Lang Hear Res* 43:879–892.

Knebel SB, Bentler RA. (1998). Comparison of two digital hearing aids. *Ear Hear 19*:280–289.

Kochkin S. (2000). MarkeTrak V: consumer satisfaction revisited. *Hear J* 53:38–55.

Kuk FK. (1990). Preferred insertion gain of hearing aids in listening and reading-aloud situations. *J Speech Hear Res* 33:520–529.

Kuk FK, Harper T, Doubek K. (1994). Preferred real-ear insertion gain on a commercial hearing aid at different speech and noise levels. *J Am Acad Audiol* 5:99–109.

Kuk FK, Lau C. (1996). Effect of hearing aid experience on preferred insertion gain selection. *J Am Acad Audiol* 7:274–281.

Kuk FK, Ludvigsen C. (1999). Variables affecting the use of prescriptive formulae to fit modern nonlinear hearing aids. *J Am Acad Audiol* 10:458–465.

Larson VD, Williams DW, Henderson WG, Luethke LE, Beck LB, Noffsinger D, Wilson RH, Dobie RA, Haskell GB, Bratt GW, Shanks JE, Stelmachowicz P, Studebaker GA, Boysen AE, Donahue A, Canalis R, Fausti SA, Rappaport BZ. (2000). Efficacy of 3 commonly used hearing aid circuits: a crossover trial. *JAMA* 284:1086–1813.

Laurence RF, Moore BCJ, Glasberg BR. (1983). A comparison of behind-the-ear high-fidelity linear hearing aids and two-channel compression aids, in the laboratory and in everyday life. *Br J Audiol* 17:31–48.

Leijon A, Eriksson-Mangold M, Bech-Karlsen A. (1984). Preferred hearing aid gain and bass-cut in relation to prescriptive fitting. *Scand Audiol* 13:157–161.

Leijon A, Lindkvist A, Ringdahl A, Israelsson B. (1990). Preferred hearing aid gain in everyday use after prescriptive fitting. *Ear Hear* 11:299–305.

Moore BCJ, Glasberg BR. (1986). A comparison of two-channel and single-channel compression hearing aids. *Audiology* 25:210–226.

Neuman AC, Bakke MH, Hellman S, Levitt H. (1995). Preferred listening levels for linear and slow-acting compression hearing aids. *Ear Hear* 16:407–416.

Scollie SD, Seewald RC, Moodie KS, Dekok K. (2000). Preferred listening levels of children who use hearing aids: comparison to prescriptive targets. *J Am Acad Audiol* 11:230–238.

Souza PE, Bishop RD. (2000). Improving audibility with nonlinear amplification for listeners with high-frequency loss. *J Am Acad Audiol* 11:214–223.

Souza PE, Turner CW. (1996). Effect of single-channel compression on temporal speech information. *J Speech Hear Res* 39:901–911

Souza PE, Turner CW. (1998). Multichannel compression, temporal cues and audibility. *J Speech Lang Hear Res* 41:315–326.

Souza PE, Turner CW. (1999). Quantifying the contribution of audibility to recognition of compression-amplified speech. *Ear Hear* 20:12–20.

Stelmachowicz P, Lewis D, Kalberer A, Creutz T. (1994). *Situational Hearing Aid Response Profile Users Manual (SHARP, v. 2.0)*. Omaha, NE: Boys Town National Research Hospital.

Stelmachowicz PG, Dalzell S, Peterson D, Kopun J, Lewis DL, Hoover BE. (1998). A comparison of threshold-based fitting strategies for nonlinear hearing aids. *Ear Hear* 19:131–138.

Tyler RS, Kuk FK. (1989). The effects of "noise suppression" hearing aids on consonant recognition in speech-babble and low-frequency noise. *Ear Hear* 10: 243–249.

Valente M, Fabry DA, Potts LG, Sandlin RE. (1998). Comparing the performance of the Widex SENSO digital hearing aid with analog hearing aids. *J Am Acad Audiol* 9:342–360.

Walden BE, Schuchman GI, Sedge RK. (1977). The reliability and validity of the comfort level method of setting hearing aid gain. *J Speech Hear Disord* 44:455–461.