Ranking Hearing Aid Input–Output Functions for Understanding Low-, Conversational-, and High-Level Speech in Multitalker Babble

King Chung  
Northwestern University

Mead C. Killion  
Etymotic Research and  
Northwestern University and  
Rush University and  
City University of New York

Laurel A. Christensen  
Etymotic Research and  
Northwestern University

Purpose: To determine the rankings of 6 input–output functions for understanding low-level, conversational, and high-level speech in multitalker babble without manipulating volume control for listeners with normal hearing, flat sensorineural hearing loss, and mildly sloping sensorineural hearing loss.

Method: Peak clipping, compression limiting, and 4 wide dynamic range compression (WDRC) input–output functions were compared in a repeated-measure design. Interactions among the compression characteristics were minimized. Speech and babble were processed and recorded at 3 input levels: 45, 65, and 90 dB sound pressure level. Speech recognition of 3 groups of listeners (n = 6/group) was tested for speech processed by each input–output function and at each input level.

Results: Input–output functions that made low-level speech audible and high-level speech less distorted by avoiding peak clipping or severe compression yielded higher speech recognition scores. These results are consistent with previous findings in the literature.

Conclusion: WDRCs with the low compression ratio region extended to a high input level or with a high compression limiting threshold were the best for speech recognition in babble when the hearing aid user cannot or does not want to manipulate the volume control. Future studies on subjective preferences of different input–output functions are needed.

KEY WORDS: hearing aid, compression, input–output function, speech recognition

One of the perceptual consequences of cochlear hearing loss is recruitment, an abnormally rapid growth in perceived loudness for a given increase in physical intensity compared with the normal rate of loudness growth (Fowler, 1928; Steinberg & Gardener, 1937). Two of the characteristics of recruitment are that the disparity between the abnormal loudness growth function and the normal loudness growth function are usually larger at low levels than at high levels and that the rate of loudness growth often differs across frequencies because people often have different degrees of hearing loss across frequencies. The implications for amplification, then, are that people with cochlear hearing loss need more gain for low-level sounds and less gain for high-level sounds and that they need different amounts of gain at different frequency regions (Skinner & Miller, 1983; Villchur, 1973). Multichannel wide dynamic range compression (WDRC) has been used extensively in hearing aids to provide such
level- and frequency-dependent amplification to minimize the need to adjust the volume control and to increase listening comfort. WDRC hearing aids typically are referred to as compression hearing aids with a low compression threshold(s).

About 30% of hearing aid users have hearing aids that do not have a volume control (Kochkin, 2005). With the growth in popularity of open-fit hearing aids, this percentage is likely to be much higher now. In addition, many users are unable to or choose not to manipulate it (e.g., young children, older users with reduced manual dexterity, users with behind-the-ear hearing aids who do not want to draw attention to their aids). Thus, the adjustment of gain in these hearing aids is largely determined by the compression circuit/algorithm and the input–output function implemented in each compression channel.

An inspection of the commercially available hearing aids revealed that a variety of input–output functions exists in WDRC hearing aids. However, there is a lack of a critical body of research comparing different types of WDRCs that would enable clinicians to make evidence-based decisions when choosing hearing aids with different types of WDRCs for users with different configurations of hearing loss. It was the focus of this study to rank linear peak clipping, compression limiting, and four WDRC input–output functions in terms of the speech recognition benefits provided to hearing aid users with flat and mildly sloping sensorineural hearing loss in situations in which no manipulation of the volume control was possible or desirable. For comparison purposes, normal-hearing listeners were included as controls.

The input–output function of a hearing aid is particularly important when the volume control of a hearing aid is not manipulated because it then almost solely controls the gain and output characteristics of the hearing aid. When displayed in hearing aid specifications, the input–output function is usually depicted as a graph with the input on the abscissa and the output on the ordinate. Several characteristics can be derived from this function: (a) the amount of gain at different input levels equals the output level minus the input level; (b) the compression ratio at an input region equals the inverse slope of the input–output function; (c) the shape of the input–output function indicates the type of signal processing strategy implemented in a certain input region (e.g., expansion, linear, compression, and curvilinear compression); (d) the knee-points of the input–output function reveal the change of one type of signal processing strategy to another (e.g., from expansion to compression); and (e) the pattern of input–output functions at different volume settings indicates whether a hearing aid has input compression or output compression.

Commercially available hearing aids exhibit a variety of input–output functions. In general, these input–output functions can be divided into three groups: linear peak clipping, linear compression limiting, and WDRC. Linear peak-clipping (or peak-clipping) hearing aids provide a constant gain for low- to mid-level inputs, and peak clipping is used at high input levels to limit the output of the hearing aid to below the discomfort level of the user (see Figure 1A). The disadvantages of peak-clipping hearing aids are the need to adjust the volume control if the sound level changes and the generation of harmonic and intermodulation distortions in response to high-level sounds (Kuk, 1996; Venema, 1998; Walker & Dillon, 1982).

Linear compression limiting (or compression limiting) hearing aids use strong compression to limit the output level of high-level sounds. The advantage of using compression limiting rather than using peak clipping is reduced harmonic and intermodulation distortions. The input–output function of compression limiting hearing aids has a linear region at low and mid-input levels and a compression region with high compression ratios (e.g., ≥8:1) at high input levels (Kuk, 1996; Venema, 1998; Walker & Dillon, 1982; see Figure 1B).

The first multichannel WDRC hearing aid simulator had a linear region at low input levels and a compression region above the compression threshold (i.e., simple WDRC; Villchur, 1973; see Figure 1C). Today, there are at least four other variations of WDRC implemented in commercially available hearing aids, and many WDRC hearing aids have three or more distinct regions in their input–output functions. One of these variations is curvilinear WDRC (see Figure 1D). Its input–output function has a linear region at very low input levels, a curvilinear region with a variable compression ratio at mid- to high input levels, and a compression limiting region at high input levels. The compression ratio in the curvilinear region may vary from 1:1 to 10:1 or more. Another variation is the high-level linear WDRC (see Figure 1E), which has linear regions at the low and high input levels and a compression region in between. The rationale of having a linear region at high input levels is that if people experience complete or partial recruitment, providing no gain or a fixed amount of gain will help normalize their loudness perception at the higher levels (Killion, 1993).

The third variation is high-level limiting WDRC (see Figure 1F) that has a linear region at low input levels, a compression region at mid-input levels, and a compression limiting region at high input levels. The compression ratios are typically between 1:1 and 3:1 for the mid-level compression region and 5:1 to 10:1 for the compression limiting region. In contrast to high-level linear WDRC, the high-level limiting WDRC is, in theory, more suitable for people with narrow dynamic ranges because it uses a compression limiting circuit to limit the output of high-level sounds. The fourth variation of WDRC was proposed by Goldberg (1996). Its input–output...
Figure 1. The input–output functions of (A) linear peak clipping, (B) linear compression limiting, (C) simple WDRC, (D) curvilinear WDRC, (E) high-level linear WDRC, (F) high-level limiting WDRC, and (G) mid-level linear WDRC. WDRC = wide dynamic range compression.
function has four regions: (a) a linear region at very low input levels, (b) a compression region at low input levels, (c) a linear region at normal conversational levels, and (d) a compression limiting region at high input levels (see Figure 1G). The rationale is that the lower compression region enhances the audibility of low-level sounds, the mid-level linear region preserves temporal envelope cues of normal conversational speech, and the compression limiting region prevents loudness discomfort.

The variety of input–output functions implemented in commercially available hearing aids is further increased by the addition of expansion (i.e., compression ratio less than 1:1) at very low input levels (typically below 35 dB SPL). Expansion was introduced to hearing aids mainly to reduce microphone noise, to reduce the pumping effect caused by the combination of low compression thresholds and syllabic compression, and/or to compensate for softness imperception as described by Buus and Florentine (2002).

Many research studies have compared the performance of hearing aids with different types of input–output functions and have yielded inconsistent findings (see Braida, Durlach, De Gennaro, Peterson, & Bustamante, 1982; Dillon, 1996; Hickson, 1994; Rintelmann, 1972; Souza, 2002, for reviews). A careful review of the literature revealed that a number of factors may have contributed to this inconsistency. One is uncontrolled or unknown differences in hearing aid performance characteristics other than those identified by the investigators or revealed by the hearing aid’s input–output functions (e.g., type of level detectors, time constants, and compression thresholds). The interactions among compression characteristics support the need to control compression characteristics other than the one under investigation. For example, the effective compression ratio is usually less than that indicated by the input–output function if long time constants are used in the compression circuit/algorithm and vice versa (Armstrong, 1993, 1997; Verschuure, Maas, Stikvoort, de Jong, Goedegebure, & Dreschler, 1996). In general, the longer the time constants, the wider the dynamic range of the output and the more the output approaches linear amplification. In addition, if the incoming signal is a rapidly varying signal (e.g., speech), the effective compression ratio can further be reduced because maximum compression (reached at the end of the attack time) is never achieved (Souza, 2002; Stone & Moore, 1992).

Besides the hearing-aid-related issues, difference in experimental design is a second factor that may make the interpretation of test results difficult. When speech recognition was tested at conversational levels or at listeners’ most comfortable levels, tested using less level-variant speech materials, or tested when the hearing aids were set to an unrealistically high gain, the speech recognition performance of listeners was usually reported to be similar among linear and WDRC hearing aids (Bille et al., 1999; Dreschler, Eberhardt, & Melk, 1984; Humes, Christensen, Bess, & Bentler, 1999; Humes, Humes, & Wilson, 2004; Kam & Wong, 1999; Larson et al., 2000; Lippmann, Braida, & Durlach, 1981; Moore, Peters, & Stone, 1999). When speech recognition was tested at low or high speech levels, the use of WDRC in hearing aids usually resulted in enhanced speech recognition, which presumably is because WDRC enhanced the audibility of low-level speech components and reduced harmonic and intermodulation distortions of high-level speech (Jenstad, Seewald, Cornelisse, & Shantz, 1999; Kam & Wong, 1999; Laurence, Moore, & Glasberg, 1983; Lippmann et al., 1981; Moore, 1987; Moore & Glasberg, 1986; Souza & Turner, 1998). Nevertheless, when speech audibility was similar or when the adjustment of the volume control was allowed during testing or during field trials, many studies found minimal differences in performance among linear and WDRC hearing aids (Humes et al., 2004; Larson et al., 2000; Shanks, Wilson, Larson, & Williams, 2002; Souza, 2002; Souza & Kitch, 2001; Souza & Turner, 1996, 1998; Walden, Surr, Cord, Edwards, & Olson, 2000). These results indicate that it is important to control the testing protocol when comparing hearing aids with different signal processing strategies and to test across a range of input levels.

The third factor making it difficult to generalize test findings on input–output functions is the listener-related variability (e.g., degree and configuration of hearing loss, previous experiences with hearing aids, and auditory dynamic range). For example, studies including listeners with mild-to-moderate hearing impairment demonstrated significantly higher speech recognition scores and perceived sound quality for compression limiting hearing aids than at high input levels (Hawkins & Naidoo, 1993; May, Dillon, & Battaglia, 1989). On the other hand, listeners with moderate or severe hearing loss preferred compression limiting hearing aids rather than linear hearing aids while listeners with profound hearing loss were divided in their preferences.

Taken together, due to the differences in hearing aid characteristics, experimental design, and listener variability, previous studies have reported inconsistent findings on hearing aid performance in terms of enhancing users’ speech recognition and subjective preferences. Therefore, when comparing the performance of different signal processing strategies, it is essential to keep the hearing aid hardware and compression characteristics—other than those characteristic(s) under investigation—constant to minimize interactions. It is also important to compare different signal processing strategies under the same testing conditions and to recruit listeners with similar audiologic characteristics.
The purpose of this study was to rank six input–output functions for speech recognition by people with flat or mildly sloping sensorineural hearing loss. Listeners with normal hearing were included as controls. Speech input levels for all listeners were set at low (45 dB), conversational (65 dB), and high (90 dB) speech levels. All speech signals were presented in a multitalker babble. The volume control setting of the amplification system was held constant across speech input level conditions. A repeated-measure design was used to minimize variability in testing protocols and listeners. The following is a detailed description of the experimental procedures.

**Method**

A two-channel experimental digital signal processing hearing aid was built to simulate six different types of input–output functions. Thus, those aspects of the hardware, software, and compression characteristics that were not manipulated as a part of this study were held constant across experimental conditions. Two compression channels were chosen for the experimental hearing aid because multiple compression channels were desirable for fitting hearing aid users with different degrees of hearing loss across frequency regions (Souza, 2002). Too many compression channels (e.g., more than four channels) may distort temporal envelope cues of speech and reduce speech recognition scores (Dreschler, 1989, 1992; Franck, van Kreveld-Bos, Dreschler, & Verschuure, 1999; Keidser & Grant, 2001; Moore & Glasberg, 1986; Stelmachowicz, Kopun, Mace, & Lewis, 1995; van Buuren, Festen, & Houtgast, 1999). As the listeners in this study had flat or only mildly sloping hearing loss, two compression channels were sufficient in providing level-dependent and frequency-dependent amplification.

The input–output functions of the experimental hearing aid were fitted to listeners with flat and sloping hearing loss using FIG6 (Etymotic Research, 1996; Gitles & Niquette, 1995; Killion & Fikret-Pasa, 1993). Institute of Electrical and Electronics Engineers (IEEE) sentences and four-talker babble were mixed at different signal-to-noise ratios (SNRs) and then fed into the experimental hearing aid at equivalent input levels of 45, 65, and 90 dB SPL, which were determined based on the input scaling of the digital chip. The output of the experimental hearing aid was recorded on compact discs and then presented monaurally to the listeners’ better ear via an ER-3A earphone. Speech recorded at 65 dB SPL was presented at each individual listener’s comfortable listening level. The presentation levels of speech processed at the 65 dB SPL input level were presented to the individual listener’s comfortable listening level. The presentation levels of speech processed at the 45 and 90 dB input levels were presented at levels as if the volume control of the experimental hearing aid was fixed. These procedures are equivalent to fitting a hearing aid to a listener, allowing the individual listener to adjust the volume control so that conversational speech is at his or her comfortable listening level, then testing the individual’s speech recognition ability at different hearing aid input levels without further volume control adjustment.

Listeners

Listeners with normal hearing and those with flat and mildly sloping sensorineural hearing loss were recruited for this study. Each listener received a hearing test in a sound-treated room to test whether his or her air and bone conduction pure-tone thresholds from 500 to 4000 Hz in octave intervals met the following audiometric selection criteria:

1. Normal hearing: 0–20 dB HL.
2. Flat sensorineural hearing loss: 50–70 dB HL. The range of thresholds for a particular listener was less than or equal to 10 dB.
3. Mildly sloping sensorineural hearing loss: 40–70 dB HL. The slope of hearing loss for a particular listener was between 15 and 30 dB.

If the above criteria were met, the speech recognition ability of listeners with normal hearing and flat hearing loss was tested using 20 IEEE sentences mixed with four-talker babble without spectral shaping (i.e., a flat frequency response). The speech recognition ability of listeners with sloping hearing loss was tested using the same sentences with high-frequency emphasis to compensate for their hearing loss. Listeners’ ability to recognize speech in babble was expressed in SNRs for 50% correct speech recognition (SNR 50s). Only listeners with SNR 50s within +/-1 dB of the group means were selected to participate in the study.

Eighty-two listeners were screened. None of them reported impaired cognitive function. Two listeners with flat hearing loss who fit the selection criteria dropped out of the study because the speech testing materials processed by some input–output functions at 90 dB SPL input level sounded too loud to them. The final group of listeners who fit the above pure-tone threshold and speech understanding criteria consisted of 10 women and 8 men aged 55–81 years (n = 6 for each group; see Table 1 for details). Power analysis using an effect size of 18%, standard deviation of difference of 12% (standard
deviation of difference was used because of the repeated-measure design), $N$ of 6, and two-tail at 95% confidence level indicated that the power of each paired $t$ test equaled 0.87.

**Experimental Digital Signal Processing (DSP) Hearing Aid**

The experimental hearing aid was built using a custom set of PC boards (Burr Brown PCM3001) that contained both a 24-bit digital signal processing (DSP) chip (DSP56303) and an 18-bit codec (Crystal-CS4215). The DSP chip had a peak processing power of 66 million calculations per second. The codec had a dynamic range of 100 dB and a sampling rate of 31250 Hz. The codec’s internal total harmonic distortion was less than 0.3%, and its frequency response was $+/-0.6$ dB from 20 Hz to 14.3 kHz. The input and output scaling of the experimental hearing aid was set at 1 volt equal to 110 dB SPL, which was also the maximum power output capability of the experimental hearing aid. Computer software programs were written to drive the experimental hearing aid. The general programming scheme is shown in Figure 2.

The experimental hearing aid had two compression channels, each with a root-mean-square (RMS) detector. The input–output function in each compression channel was defined by mathematical equations specifying the relation between the input and the output from equivalent input levels of 0–120 dB SPL. The input–output functions

<table>
<thead>
<tr>
<th>Listener Group</th>
<th>Age (in years)</th>
<th>$n$</th>
<th>Hearing Thresholds 0.5−4.0 kHz</th>
<th>SNR 50s</th>
<th>SNRs of Speech and Babble (dB)</th>
</tr>
</thead>
<tbody>
<tr>
<td>NH</td>
<td>60.4</td>
<td>3.9</td>
<td>6</td>
<td>4−8</td>
<td>3.8−6.1</td>
</tr>
<tr>
<td>FLAT</td>
<td>72.8</td>
<td>4.3</td>
<td>6</td>
<td>56.7−60.8</td>
<td>4.1−7.4</td>
</tr>
<tr>
<td>SLOPE</td>
<td>72.7</td>
<td>8.5</td>
<td>6</td>
<td>42−67</td>
<td>2.6−5.8</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Listener Group</th>
<th>M (in years)</th>
<th>SD</th>
<th>Hearing Thresholds 0.5−4.0 kHz</th>
<th>SD</th>
<th>SNR 50s</th>
<th>Range (dB)</th>
<th>SNRs of Speech and Babble (dB)</th>
</tr>
</thead>
<tbody>
<tr>
<td>NH</td>
<td>60.4</td>
<td>3.9</td>
<td>6</td>
<td>4−8</td>
<td>3.8−6.1</td>
<td>4.0−5.4</td>
<td>+7, +2</td>
</tr>
<tr>
<td>FLAT</td>
<td>72.8</td>
<td>4.3</td>
<td>6</td>
<td>56.7−60.8</td>
<td>4.1−7.4</td>
<td>9.0−9.8</td>
<td>+12, +7</td>
</tr>
<tr>
<td>SLOPE</td>
<td>72.7</td>
<td>8.5</td>
<td>6</td>
<td>42−67</td>
<td>2.6−5.8</td>
<td>6.3−7.8</td>
<td>+9, +4</td>
</tr>
</tbody>
</table>

**Table 1.** The mean age, hearing thresholds, signal-to-noise ratio (SNR) for 50% correct speech recognition, and SNRs of the recorded speech test materials for listeners with normal hearing (NH), flat sensorineural hearing loss (FLAT), and sloping sensorineural hearing loss (SLOPE).

Figure 2. The schematics of the two-channel experimental digital signal processing hearing aid. RMS = root mean square.
were implemented in the form of lookup tables in the experimental hearing aid. The type of the input–output functions in the two channels was kept the same in the experimental hearing aid because most high-performance hearing aids have the same type of input–output functions across compression channels. The crossover frequency of the compression channel was set at 1500 Hz, and the slopes of the filters dividing the two compression channels were 18 dB/octave. The experimental hearing aid was implemented with variable time constants. The attack time was less than 10 ms. The release time varied from 10 to 600 ms when tested with signals with 1- to 1,000-ms durations, respectively (Figure 3). A special algorithm was written to keep these values similar among the input–output functions and among the compression regions with different compression ratios.

After examining the input–output functions of commercially available hearing aids, the experimental hearing aid was programmed to simulate six commonly used input–output functions with the following characteristics:

1. Linear peak clipping (LinPC): linear with maximum power output (peak-clipping level) at 105 dB SPL.
3. Curvilinear WDRC (Curvi): linear for input levels below 37 dB SPL, curvilinear compression for input levels between 37 and 85 dB SPL, and compression limiting for levels above 85 dB SPL.
4. High-level-linear WDRC (LWDC): linear for input levels below 45 dB SPL and above 90 dB SPL, and compression for input levels between 45 and 90 dB SPL.
5. High-level-limiting WDRC (FDRC): linear for input levels below 45 dB SPL, compression for input levels between 45 and 85 dB SPL, and compression limiting for input levels above 85 dB SPL.
6. Mid-level-linear WDRC (MLWDC): linear for input levels below 30 dB SPL and between 55 and 75 dB SPL, compression for input levels between 30 and 55 dB SPL, and compression limiting for input levels above 75 dB SPL.

Simple WDRC was not included because it is not commonly used in high-performance hearing aids.

**Hearing Aid Fitting**

The experimental hearing aid was fitted to each group of listeners using the FIG6 Hearing Aid Fitting Prescription (Etymotic Research, 1996). FIG6 recommends target gains at input levels of 40, 65, and 95 dB SPL based on the differences between the average normal loudness growth functions of people with normal hearing and people with hearing impairment. It prescribes more gain for low-level inputs than for high-level inputs. As the average thresholds for listeners with flat hearing loss were 56.7 dB HL at 500 and 60.8 dB HL at 4000 Hz, the low- and high-frequency channels of the experimental hearing aid were programmed to have the same gains as if the listeners had a flat hearing loss of 60 dB HL (see Figure 4). Listeners with sloping hearing loss had average thresholds of 41.6 dB HL at 500 Hz and 66.7 dB HL...
at 4000 Hz. The low- and high-frequency channels of the experimental hearing aid were programmed to fit hearing losses of 45 dB HL and 70 dB HL, respectively (see Figure 5). The maximum power output of LinPC was set at 105 dB SPL for all listener groups.

The gain of each input–output function for an input level of 65 dB SPL was always matched to the FIG6 target gain to ensure a comfortable listening level for normal conversations. The gains at input levels of 40 and 95 dB SPL were programmed to approximate the target gains as closely as possible. However, for some input–output functions, it was not feasible to achieve the target gains at all three input levels (e.g., LinPC and LinCL provided fixed gains for low- and mid-level inputs). It could only meet the target gains at either 40 or 65 dB SPL. In these cases, the target gains at 65 dB SPL were met. Thus, the achieved gains at 40 dB SPL were much lower than the target gains for low-level inputs.

Figure 5. The target gains at 40, 65, and 95 dB SPL input recommended by FIG6 and the input–output functions of the low- and high-frequency channels of the six input–output functions fitted to listeners with sloping hearing loss.

In addition to keeping the gain at 65 dB SPL input constant, the compression ratios for the six input–output functions at similar input regions were also kept constant (see Table 2). The compression ratios of the mid-level compressors were chosen based on two considerations: (a) the compression ratios should not exceed 3.1:1 because previous studies have shown that higher compression ratios often degrade speech recognition and (b) the selected compression ratios should allow the gains of the WDRCs at 40- and 95-dB SPL input levels to meet or be reasonably close to the corresponding target gains. Therefore, the compression ratio in the low- and high-frequency channels was chosen to be 2.5:1 when input–output functions were fitted to listeners with flat hearing loss; for listeners with sloping hearing loss, compression ratios were chosen to be 1.8:1 and 3.1:1 in the low- and high-frequency channels, respectively.

If conflicts existed between meeting the target gains at 40 or 95 dB SPL and conforming to the compression ratio, the latter was given precedence. For example, when LWDR was fitted to listeners with flat hearing loss, the target gains for 40, 65, and 95 dB SPL inputs were 40, 24, and 7 dB, respectively. If LWDR were programmed to meet all the target gains, the compression ratio for the mid-level compressor would have been 3.1:1. To keep the compression ratio of the mid-level compressor constant at 2.5:1 for all six input–output functions, the gain of LWDCs for 95 dB input was increased to 9 dB.

If it was called for in the input–output function, the compression ratio of the high-level compressor was set to 8:1. This is because most of the high-level compressors used for compression limiting purposes in commercially available hearing aids have compression ratios of between 5:1 and 10:1.

**Verification of Compression Characteristics**

In the developmental stage, the experimental hearing aid went through rigorous cycles of testing, cross-checking, fine tuning, and verification. Its frequency responses, input–output functions, time constants, and total harmonic distortions were tested and verified by using at least two of the following three testing methods or by conducting two different tests using the same testing method: (a) custom MATLAB testing programs,
(b) electroacoustic analysis in a Frye 6500 Hearing Aid Analyzer, and (c) manual measurement methods (i.e., feeding electrical signals to the input of the experimental DSP hearing aid and measuring the output using a voltmeter; Chung, 2001). In general, the custom MATLAB programs provided the cleanest test results because electric signals generated by the computer were directly fed into the experimental hearing aid and the hearing aid output was directly measured by the computer. Thus, a minimum amount of noise interference was present in the signal processing path.

The algorithms of the experimental hearing aid were fine tuned until the measured values matched the expected values. For input–output function measurements, the expected values were the values used in the lookup tables for each type of input–output function fitted to the two groups of listeners with hearing loss. For frequency responses, the expected values were calculated from 100 to 10000 Hz based on the input–output functions and the slope of the filters separating the two compression channels. The harmonic distortions of the experimental hearing aid should be as low as possible, with the

<table>
<thead>
<tr>
<th>Level</th>
<th>LinPC</th>
<th>LinCL</th>
<th>Curvi</th>
<th>LWDRC</th>
<th>FDRC</th>
<th>MLWDRC</th>
</tr>
</thead>
<tbody>
<tr>
<td>High level</td>
<td>8:1</td>
<td>8:1</td>
<td>1:1</td>
<td>8:1</td>
<td>8:1</td>
<td>8:1</td>
</tr>
<tr>
<td>Mid-level</td>
<td>1:1</td>
<td>1:1</td>
<td>(1.4–7.2)</td>
<td>2.5:1</td>
<td>2.5:1</td>
<td>1:1</td>
</tr>
<tr>
<td>Low level</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>2.5:1</td>
</tr>
<tr>
<td>Very low level</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
</tr>
</tbody>
</table>

**Sloping hearing loss (LF channel)**

<table>
<thead>
<tr>
<th>Level</th>
<th>LinPC</th>
<th>LinCL</th>
<th>Curvi</th>
<th>LWDRC</th>
<th>FDRC</th>
<th>MLWDRC</th>
</tr>
</thead>
<tbody>
<tr>
<td>High level</td>
<td>8:1</td>
<td>8:1</td>
<td>1:1</td>
<td>8:1</td>
<td>8:1</td>
<td>8:1</td>
</tr>
<tr>
<td>Mid-level</td>
<td>1:1</td>
<td>1:1</td>
<td>(1.4–7.2)</td>
<td>1.8:1</td>
<td>1.8:1</td>
<td>1.8:1</td>
</tr>
<tr>
<td>Low level</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
</tr>
<tr>
<td>Very low level</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
</tr>
</tbody>
</table>

**Sloping hearing loss (HF channel)**

<table>
<thead>
<tr>
<th>Level</th>
<th>LinPC</th>
<th>LinCL</th>
<th>Curvi</th>
<th>LWDRC</th>
<th>FDRC</th>
<th>MLWDRC</th>
</tr>
</thead>
<tbody>
<tr>
<td>High level</td>
<td>8:1</td>
<td>8:1</td>
<td>1:1</td>
<td>8:1</td>
<td>8:1</td>
<td>8:1</td>
</tr>
<tr>
<td>Mid-level</td>
<td>1:1</td>
<td>1:1</td>
<td>(1.4–7.2)</td>
<td>3.1:1</td>
<td>3.1:1</td>
<td>3.1:1</td>
</tr>
<tr>
<td>Low level</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
</tr>
<tr>
<td>Very low level</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
<td>1:1</td>
</tr>
</tbody>
</table>

**Note.** WDRC = wide dynamic range compression; LinPC = linear peak clipping; LinCL = linear compression limiting; Curvi = curvilinear WDRC; LWDRC = high-level-linear WDRC; FDRC = high-level-limiting WDRC; MLWDRC = mid-level-linear WDRC. Dashes indicate that data are not applicable.

*Indicates level at which clipping starts. Compression ratio varied from 1.4:1 to 7.2:1.*
exception of LinPC at high input levels when it went into saturation.

In the final verification process, the functions of the experimental hearing aid were tested when it was programmed to each of the six input–output functions. The frequency responses and the input–output functions of the experimental hearing aid were cross-checked and verified by custom MATLAB programs, manual measurements, and electroacoustic measurements. When tested with the custom MATLAB programs, the frequency responses were within +/-3 dB of the expected values from 10 to 10,000 Hz from equivalent input levels of 45–105 dB SPL in 10-dB steps. The input–output functions were +/-3 dB within the expected values for all six input–output functions for both groups of listeners.

Harmonic distortion of the experimental hearing aid was measured using the American National Standards Institute (ANSI) 1987 testing program and pure-tone sweeps. The ANSI measurement indicated that the total harmonic distortions were less than 3.8% at 500, 800, and 1600 Hz at 70, 70, and 65 dB SPL for all six input–output functions, respectively. For input–output functions other than LinPC, the total harmonic distortions measured using pure-tone sweeps were less than 3% (with a mean of 0.5%) from 50 to 100 dB SPL. For LinPC, the harmonic distortions were less than 3% for pure-tone sweeps at levels below saturation and up to 13.4% when the hearing aid went into saturation.

The time constants of the experimental hearing aid were measured by the time constant estimation with variable frequency (TCEVF) program, manual measurement, and the ANSI 1996 standard test in the Frye 6500 Hearing Aid Analyzer. Manual measurements were used to spot-check the results of the TCEVF program. The measurements made in the Frye 6500 were used to document the time constants estimated under standardized conditions. The TCEVF was written and modified by Joan Miller. It allowed the investigators to define the frequency, level, and duration of the testing signal and the number of dB approaching the steady-state level for time constant calculations. In the TCEVF program, the attack time was defined as the duration between the time when the input level of a pure tone abruptly increased 25 dB and the time when the output level was stabilized to within 3 dB of the steady-state value. The release time was defined as the duration between the time when the input level of a pure tone abruptly decreased 25 dB and the time when the output level stabilized to within 2 dB of the steady-state value. The time constants of all the compression regions in each input–output function were measured using a 500-Hz tone and a 2000-Hz tone.

As compressors were implemented at different input regions in the six input–output functions, the presentation levels of the test tones were varied so that the low- and high-level test tones fell within the compression region. When testing the time constants of the mid-level compressors (i.e., Curvi, LWDRC, and FDRC), the test tone varied from equivalent input levels of 55–80–55 dB SPL. If the compression region was below or above these levels (e.g., the low-level compressor of MLWDRC or the high-level compressor of FDRC and LinCL), the presentation levels of the testing tones were reduced or increased, but the level differences were always kept at 25 dB. If the compression region had a dynamic range narrower than 25 dB, the knee-points of the compressor were temporarily shifted up or down so that the levels of the testing tones could fall within the compression region.

For example, FDRC had two compression regions: a mid-level compressor acts on inputs between 45 and 85 dB SPL, and a high-level compressor acts on inputs above 85 dB SPL. The attack and release times of the mid-level compressor were tested using a pure-tone varying from equivalent input levels of 55–80–55 dB. To test the time constants of the high-level compressor, the compression threshold of the higher compression region was temporarily lowered to 70 dB and the testing tone was presented at equivalent levels of 75–100–75 dB SPL. Accounting for all compression regions of the five input–output functions, the attack times of the two frequency channels were always less than 10 ms. The averages and standard deviations of the release times tested using signals with variable durations are shown in Figure 3.

**Recording of the Speech Testing Materials**

After the proper functions of the experimental hearing aid were checked and verified, the IEEE sentences were time locked with the four-talker babble and recorded in the left and right channel of digital audiotapes. This synchronization was performed to ensure that any disparities in the processed sentences or in listeners’ performance were due to differences in input–output function and not due to variations in the instantaneous change of SNR of the speech and babble. During the recording of the testing materials, the level of the IEEE sentences was adjusted so that the speech peaks were presented to the experimental hearing aid at equivalent input levels of 45, 65, and 90 dB SPL. As 1 volt equals 110 dB SPL at the input of the experimental DSP hearing aid (i.e., the input scaling), the calibration tone of the IEEE sentences was presented at 0.00056 V (45 dB), 0.0056 V (65 dB), and 0.1 V (90 dB). The level of the babble was adjusted relative to the level of speech to reflect the SNR needed for the experimental condition. The IEEE sentences were spoken by a female speaker in standard American English.

To test listeners’ speech recognition scores at three input levels, 720 IEEE sentences were equally divided into three sets. Each set of sentences was processed by each of the six input–output functions at one input level. Half of...
the sentences within each set were recorded at an SNR (high SNR) that was 2.5 dB above the group mean of SNR 50 obtained in the participant selection process for each group of listeners. The other half were recorded at a SNR (low SNR) of 2.5 dB below the group mean. For example, the group mean of SNR 50 for listeners with flat hearing loss was 9.5 dB. The speech and babble of the testing materials were recorded at SNRs of 12 dB and 7 dB at the input of the experimental hearing aid. This paradigm was used in order to reduce floor-and-ceiling effects. The test materials for listeners with normal hearing were made through the experimental hearing aid fitted to listeners with flat hearing loss but recorded at lower SNRs (i.e., 7 dB and 2 dB), as listeners with normal hearing have better speech recognition abilities in noise than do listeners with flat hearing loss (see Table 1 for details).

The sentences and babble processed by the six input–output functions were recorded on digital audiotapes using a Mackie 16 Channel Mic/Line Mixer (CR1604-VLZ). As the frequency response of ER-3A insert earphones has a high-frequency roll-off above 4000 Hz, a high-pass filter and an ER11-80D amplifier were added to the circuit so that the real-ear frequency response of the ER-3A insert earphones was equalized to be relatively flat up to 6 kHz.

### Data Collection

Prior to testing speech recognition, loudness growth functions for each listener were estimated using the Independent Hearing Aid Fitting Forum (IHAFF) procedures (Valente & Van Vliet, 1997). The signal was a sentence processed by LinPC at 65 dB SPL. The average intensity for a comfortable rating (i.e., 4 on the IHAFF scale) was then determined by calculating the median of three ascending runs. Subsequently, the testing materials processed at 65 dB SPL input were presented at each listener’s comfortable listening level (i.e., the 4 level). The averages of the 4 level were 60, 81, and 82 dB HL for listeners with normal hearing, flat hearing loss, and sloping hearing loss, respectively. The sentences recorded at 45 and 90 dB SPL input levels were presented at audiometer dial levels as if the volume control of the experimental hearing aid was fixed.

During the data collection process, each listener listened to 20 sentences at a low SNR, and 20 sentences were presented at a high SNR processed by each input–output function at each input level. He or she was instructed to repeat as many words as possible after listening to a sentence. He or she was encouraged to guess if not sure. The examiner wrote down the listener’s responses for offline scoring. Each IEEE sentence had five keywords with 1 point given for each correctly repeated keyword. If a listener did not repeat a keyword correctly but repeated more than half of the total phonemes in the keyword, a half point was given. The speech recognition score was then calculated by counting the number of points a listener received for each input–output function. The testing was carried out over 3 to 4 days. At the end of the study, each listener had listened to all 720 IEEE sentences (i.e., 20 sentences × 2 SNRs × 6 input–output functions × 3 input levels). The presentation order of the input–output function and input levels was counterbalanced, and the assignment of sentence lists to each input–output function was also counterbalanced within each listener group. Listeners were not blinded to the experimental conditions during testing.

### Results

The mean speech recognition scores obtained by the three listener groups at the three input levels are

---

**Table 3.** The RMS presentation levels of sentences (in babble) processed by the six input–output functions and the output dynamic range of the sentences processed at high SNR ratios and equivalent input levels of 45, 65, and 90 dB SPL.

<table>
<thead>
<tr>
<th>Function</th>
<th>90</th>
<th>65</th>
<th>45</th>
<th>Dynamic Range</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Normal hearing</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>LinPC</td>
<td>79.9</td>
<td>60.0</td>
<td>36.8</td>
<td>43.1</td>
</tr>
<tr>
<td>LinCL</td>
<td>79.9</td>
<td>60.0</td>
<td>36.8</td>
<td>43.1</td>
</tr>
<tr>
<td>Curvi</td>
<td>72.5</td>
<td>63.7</td>
<td>51.5</td>
<td>20.9</td>
</tr>
<tr>
<td>LWDRC</td>
<td>74.0</td>
<td>63.9</td>
<td>51.8</td>
<td>22.2</td>
</tr>
<tr>
<td>FDRC</td>
<td>74.0</td>
<td>64.0</td>
<td>51.9</td>
<td>22.2</td>
</tr>
<tr>
<td>MLWDRC</td>
<td>78.4</td>
<td>60.8</td>
<td>50.0</td>
<td>28.4</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th><strong>Flat hearing loss</strong></th>
</tr>
</thead>
<tbody>
<tr>
<td>LinPC</td>
</tr>
<tr>
<td>LinCL</td>
</tr>
<tr>
<td>Curvi</td>
</tr>
<tr>
<td>LWDRC</td>
</tr>
<tr>
<td>FDRC</td>
</tr>
<tr>
<td>MLWDRC</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th><strong>Sloping hearing loss</strong></th>
</tr>
</thead>
<tbody>
<tr>
<td>LinPC</td>
</tr>
<tr>
<td>LinCL</td>
</tr>
<tr>
<td>Curvi</td>
</tr>
<tr>
<td>LWDRC</td>
</tr>
<tr>
<td>FDRC</td>
</tr>
<tr>
<td>MLWDRC</td>
</tr>
</tbody>
</table>
shown in Figure 6. The graphs showed that the SNRs that were chosen indeed avoided the ceiling and floor effects. Thus, the raw scores were used in the following analysis.

As the primary interest of this study was to differentiate the effects of input–output functions on speech recognition, a 2 (SNR) × 6 (input–output function) analysis of variance (ANOVA) with repeated measure on the input–output function variable was conducted at each input level for each listener group. A significant main effect of SNR \((p < .05)\) was observed for all listener groups and at all three input levels. In general, the high-SNR conditions generated higher speech recognition scores than their corresponding low-SNR conditions.

A significant input–output function effect was obtained at the 90 dB SPL input level for all listener groups and at the 45 dB SPL input level for listeners with hearing loss at both high and low SNRs \((p < .05)\). No significant input–output function effect was observed at 65 dB SPL input level for any listener group at either SNR. In addition, there was no significant interaction effect between SNR and input–output function.

Post hoc tests were conducted on the sets of data that showed significant input–output function effect. Bonferroni–Dunn paired comparisons were performed within each subject group and within the same SNR conditions to identify the input–output functions that generated significantly different speech recognition scores. The 5% significance level was adjusted to \(p < .0033\) to account for multiple paired comparisons. A summary of the results is tabulated in Table 4.

For listeners with normal hearing, the only significant input–output function effect was observed at the 90 dB SPL input level. Post hoc Bonferroni–Dunn pairwise comparisons indicated that listeners’ speech recognition scores obtained using LinCL, Curvi, LWDRC, FDRC, and MLWDRC were significantly better than those obtained with LinPC in both SNR conditions \((p < .0004)\).

For listeners with flat hearing loss, significant input–output function effects were observed at input levels of 45 dB and 90 dB SPL in both high (12 dB) and low (7 dB) SNR conditions. Pairwise comparisons revealed that the WDRCs yielded significantly higher speech recognition scores than LinPC and LinCL at 45 dB SPL in both SNR conditions \((p < .0006)\). At the input level of 90 dB SPL, all other input–output functions obtained significantly higher speech recognition scores than LinPC in both SNR conditions \((p < .0008)\). LWDRC also yielded higher scores than LinCL and Curvi in low-SNR conditions \((p < .0032\) and \(p < .0008\), respectively).

For listeners with sloping hearing loss, the WDRCs yielded significantly higher speech recognition scores than LinPC and LinCL at both high (9 dB) and low (4 dB) SNRs at 45 dB SPL input level \((p < .0001)\). At the input level of 90 dB SPL, LWDRC and FDRC were found to be significantly better than LinPC in high-SNR conditions \((p < .0013\) and \(p < .0009\), respectively). In addition,
LWDRC yielded significantly higher scores than LinPC, LinCL, and MLWDRC in low-SNR conditions \((p < .0001, p < .0018, \text{and } p < .0003, \text{respectively})\).

**Discussion**

In this study, speech recognition scores were obtained when listeners listened to speech processed by six input–output functions at input levels of 45, 65, and 90 dB SPL. The testing materials were presented at levels as if the volume control of the two-channel experimental hearing aid was fixed. Three groups of homogeneous listeners with normal hearing and those with mild to moderately severe flat or sloping hearing loss participated in the study. Therefore, the following discussion is pertinent to hearing aid users with similar degrees and configurations of hearing loss and for hearing aids without volume controls or for hearing aid users who are not likely to adjust the volume control.

Clinically, the results of this study suggest that the choice of input–output function for a hearing aid user depends on the hearing aid user’s degree and configuration of hearing loss. For listeners with flat hearing loss, speech recognition scores in speech babble while listening to the six input-output functions increased from...
WDRCs provided 12.2 conditions produced significantly higher speech recognition of audibility for low-level speech. First, the WDRC the recognition of high-level speech. This distortion can be generated significant distortion at 90 dB SPL. Although LinPC also yielded lower speech recognition scores than the WDRCs at 45 dB SPL input due to audibility issues, it obtained significantly better scores than LinPC at 90 dB SPL because it avoided peak clipping by utilizing compression. The WDRCs, as a group, were more appropriate than LinPC and LinCL for listeners with flat hearing loss. No significant differences were observed among these WDRCs in the high-SNR conditions at any input levels. LWDRC, however, yielded higher speech recognition scores than Curvi and LinCL at 90 dB SPL in the low-SNR conditions at 90 dB SPL. Thus, LWDRC appeared to be more appropriate than Curvi for listeners with flat hearing loss. The lower ranking of the Curvi compared with other WDRCs probably occurred because its high compression ratio region started at a relatively low input level and the severe compression affected speech recognition of high-level speech.

For listeners with sloping hearing loss, the ranking of the input–output function progresses from the worst LinPC, to LinCL, to Curvi and MLWDRC, to FDRC, and to LWDR. Similar to the results from listeners with flat hearing loss, listeners with sloping hearing loss received significantly lower speech recognition scores when listening to LinPC and LinCL than to WDRCs at 45 dB SPL. Yet the only significantly better scores observed at 90 dB SPL input occurred when LWDRC was compared with LinPC, LinCL, and MLWDRC in the low-SNR conditions and when LWDRC and FDRC were compared with LinPC in the high-SNR conditions. In other words, consistent benefits of WDRCs over LinPC and LinCL were shown only at low-level input and LWDRC seemed to be the best input–output function of the six included in this study for listeners with sloping hearing loss.

When the general trends are analyzed across listener groups, the results of this study provided additional support to previous findings that (a) audibility is essential for understanding low-level speech; (b) compression with a compression ratio of less than 3:1 does not degrade or enhance speech recognition at conversational levels; and (c) the amount of distortion, either from peak clipping or from severe compression, affects the recognition of high-level speech.

Several findings of this study support the importance of audibility for low-level speech. First, the WDRC conditions produced significantly higher speech recognition scores for listeners with hearing loss because the WDRCs provided 12.2–14.2 dB more gain than LinPC and LinCL for speech at the 45 dB SPL input (see Table 3). Many listeners with hearing impairment reported that they could not hear the speech when they listened to the LinPC and LinCL conditions, but they were able to hear the speech when they listened to the WDRC conditions. Second, when the audibility of the speech was similar, either among WDRC conditions at 45 dB SPL or among all input–output functions at 65 dB SPL, listeners with hearing loss obtained similar scores. Third, listeners with normal hearing obtained similar SNR 50s for all six input–output functions at input levels of 45 and 65 dB SPL because all of the speech components were audible to them. These findings are consistent with previous studies reporting that WDRCs enhance audibility and improve the recognition of low-level speech (Braida et al., 1982; Dillon, 1996; Hickson, 1994; Jenstad et al., 1999; Kam & Wong, 1999; Laurence et al., 1983; Lippmann et al., 1981; Moore, 1987; Moore & Glasberg, 1986; Rintelmann, 1972; Souza, 2002; Souza & Turner, 1998). A logical deduction is that multiple WDRCs increase the audibility of low-level sounds, and the exact shape of the WDRC is less important.

Another general conclusion that also supports previous findings is that compressors with compression ratios of less than or equal to 3:1 do not degrade or enhance speech recognition when all of the speech components are audible. In our study, the compression ratios of the low- and mid-level compressors were set between 1.8:1 and 3:1:1 for all WDRCs. When speech was presented to the experimental hearing aid at 65 dB SPL, speech was processed either at the linear region (LinPC and LinCL) or at the compression region with compression ratios of less than 3:1. The lack of significant difference in speech recognition scores across listener groups and experimental conditions at 65 dB SPL input level is consistent with the results of previous studies: Compressors with low compression ratios do not degrade or enhance speech recognition if the audibility of the speech components is similar across conditions (Dillon, 1996; Jenstad et al., 1999; Souza & Turner, 1999; van Buuren, Festen, & Houtgast, 1996; van Harten-de Bruijn, van Kreveld-Bos, Dreschler, & Verschuure, 1997).

A third general conclusion drawn from the results of this study is that the amount of distortion affects the recognition of high-level speech. This distortion can be either from severe compression (i.e., compressors with high compression ratios) or in the form of harmonic and intermodulation distortions that are generated in the peak-clipping process. The negative effects of peak clipping on speech recognition have been well documented in previous studies (Braida et al., 1982; Dillon, 1996; Dreschler, 1988; Hawkins & Naidoo, 1993; Hickson, 1994; Kam & Wong, 1999; Kuk, 1996; Larson et al., 2000; Naidoo & Hawkins, 1997; Rintelmann, 1972; Souza, 2002; Storey, Dillon, Yeend, & Wigney, 1998; Walker &
Dillon, 1982). In this study, listeners with normal hearing and flat hearing loss obtained an average of 17%–45% higher speech recognition scores with LinCL and the WDRCs than LinPC at 90 dB SPL. This indicates that high-level compressors with compression limiting (LinCL, Curvi, FDRC, and MLWDRC) or linear amplification without peak clipping (LWDRC) are better than peak clipping for understanding high-level speech.

The negative effect of peak clipping on speech recognition is also observed in listeners with sloping hearing loss at 90 dB SPL input level but in a more subtle manner. Compared with LinPC, only LWDRC and FDRC yielded significantly higher scores at high (9 dB) SNR, and only LWDRC yielded significantly higher scores at low (4 dB) SNR. The scores obtained by LinCL, Curvi, and MLWDRC are not significantly different from those of LinPC. These results seem to be counterintuitive to our knowledge that compared with peak clipping, compression limiting improves speech recognition for high-level speech. An inspection of the input–output functions revealed that LinPC started to peak clip when the overall input level in either frequency channel was 81 dB SPL or higher for listeners with flat hearing loss and normal hearing. For listeners with sloping hearing loss, on the other hand, peak clipping did not occur until the input level exceeded 90 dB SPL in the low-frequency channel or 75 dB SPL in the high-frequency channel.

To investigate the effect of having different amounts of gain and saturation levels in the experimental hearing aid, unprocessed IEEE sentences were concatenated and mixed with the four-talker babble at SNRs of +4, +7, +9, and +12 dB to simulate the low- and high-SNR inputs of the two groups of listeners. Then these waves were filtered by a low- and a high-pass filter with similar characteristics as those implemented in the experimental hearing aid (i.e., cutoff frequency at 1500 Hz and a slope of 18 dB per octave). The average RMS levels of the unfiltered and the filtered waves were measured. The results showed that the low-pass-filtered waves had RMS levels of approximately 0.4 dB below those of the unfiltered waves, and the high-pass-filtered waves had RMS levels of approximately 12 dB below those of the low-pass-filtered waves across SNRs. This means that the sound pressure level detected in the high-frequency channel was approximately 12 dB lower than that in the low-frequency channel.

When the speech peaks were presented at an equivalent input level of 90 dB SPL to the experimental hearing aid, the speech peaks were presented at approximately 89.6 and 77.6 dB SPL in the low- and high-frequency channels, respectively. This means that the speech peaks in the low-frequency channel were roughly 8.6 dB into saturation for listeners with flat hearing loss, whereas the speech peaks were just 2.6 dB into saturation in the high-frequency channel for listeners with sloping hearing loss. Consequently, the speech processed by LinPC was less distorted for listeners with sloping hearing loss than the speech processed for listeners with flat hearing loss and normal hearing. Thus, many input–output functions with compression at high-level regions failed to yield significantly better speech recognition scores than LinPC for listeners with sloping hearing loss but did yield better scores for listeners with flat hearing loss. These results are consistent with the previous findings that the higher the amount of saturation, the more negative the effects on understanding high-level speech (Dawson et al., 1990; Hawkins & Naidoo, 1993).

The differences in significance patterns between the listener groups with hearing loss among the six input–output functions suggest that hearing aid users with flat hearing loss may experience more detrimental effects with peak clipping hearing aids than users with sloping hearing loss, even though they have similar pure-tone averages. This disparity indicates that the effect of input–output function on speech recognition is determined by the interaction among the gain needed at different frequency regions, the maximum power output of the hearing aid, and the energy distribution of speech. These factors determine the amount of saturation when the hearing aid is in operation and affect the hearing aid user’s ability to recognize high-level speech.

A second possible source of distortion for high-level speech was severe compression. The significance patterns in Table 4 indicate that there were no significant differences among the WDRCs and LinCL in most of the 90 dB SPL conditions among all listener groups. When a significant difference was detected, however, LWDRC was the one to yield higher scores than other input–output functions. To understand the significance of this pattern, the knee-points of the input–output functions and the presentation levels of the speech components at low- and high-frequency regions were examined.

As mentioned earlier, when speech peaks were presented at 90 dB SPL, the levels detected at the low- and high-frequency channels were approximately 89.6 and 77.6 dB SPL, respectively. As the range of the mid-level compressor of LWDRC extended to 90 dB SPL, the speech sounds were processed at compression ratios of lower than 3:1 in both channels. For other input–output functions, the low-frequency speech sounds were processed by the high-level compression limiters that have compression thresholds of 85 dB or lower and a compression ratio of 8:1 (see Table 2 for details). Previous studies have reported that severe compression induced by compressors with compression ratios above 5:1 distorts the temporal envelopes of signals, reduces the temporal contrasts between consonants and vowels,
and thus degrades speech recognition (Boike & Souza, 2000; Boothroyd, Springer, Smith, & Schulman, 1988; Bustamante & Braida, 1987; Plomp, 1994; Souza, 2002; van Tasell & Trine, 1996). In this study, significant differences observed between LWDRD and other input–output functions with high-level compression limitors were likely due to the presence of severe compression. Apparently, the negative effects of such distortion were more prominent at low SNRs than at high SNRs because the temporal contrasts between speech and noise were already reduced at low SNRs.

It is interesting to note that no significant differences were found between LWDRD and input–output functions other than LinPC for listeners with normal hearing. This finding is consistent with that of Hornsby and Ricketts (2001). The speech recognition ability of listeners with normal hearing seems to be unaffected by severe compression at high levels.

It is also worth noting that the superior performance obtained by LWDRD in this study is not directly related to its linear amplification above 90 dB SPL because all of the speech components were presented at or below 90 dB SPL. The determining factor was that speech was processed at a compression ratio lower than the compression ratio used in the compression limiting region. LWDRD was essentially acting like a simple WDRC in this study. The findings of this study suggest that the best input–output functions increase the audibility of low-level sounds and generate minimum distortions, either by peak clipping or severe compression, at high levels. The implications of this for other types of WDRCs are that the compression thresholds of the compression limiting region should be set as high as possible to avoid the negative effects of severe compression on speech recognition.

A cautionary implication for hearing aid fitting is that high-performance hearing aids may use compression limiting to limit the output of the hearing aid in order to avoid digital clipping. The controls of this compression limiting algorithm are often not shown on the gain setting screen in the fitting software but, instead, are controlled by a combination of settings such as maximum power output, reserve gain, or under other names with similar functions. As the maximum power output is set at a lower and lower number, or as the reserve gain is set at a higher and higher value, the compression threshold of the compression limiting region becomes lower and lower. This can potentially affect the hearing aid user’s speech understanding of high- or conversational-level speech. Therefore, it is recommended that audiologists routinely check these settings to make sure that the compression threshold of the high-level compressor is set at the highest value possible without the hearing aid user complaining of discomfort caused by loud sounds.

A secondary finding of this study is that Curvi may be the most suitable input–output function for hearing aid users with flat hearing loss and very narrow auditory dynamic ranges. Two listeners with flat hearing loss had auditory dynamic ranges of approximately 30–35 dB. When speech at 65 dB SPL was presented to them at their comfortable level, they complained that the speech processed at 90 dB SPL was too loud. The only condition they could tolerate at 90 dB SPL input was Curvi. Subsequently, they opted to drop out of the study. The measurement results in Table 3 suggest that Curvi has the least output dynamic range among the input–output functions for listeners with flat hearing loss. Apparently, Curvi, LWDRD, and FDRC have the smallest output dynamic range for listeners with sloping hearing loss. More future studies are needed to determine if this finding holds true for a larger number of listeners with narrow dynamic ranges.

The findings of this study indicate that the audibility of low-level sounds and minimum distortion at high levels are the determining factors for a good hearing aid if the volume control of the hearing aid is not adjusted. Although the experimental hearing aid was fitted using a prescription based on the loudness normalization philosophy, it is likely that these findings can be generalized to cases in which hearing aids are fitted by the speech recognition maximization philosophy (e.g., NAL–NL1). The speech recognition maximization philosophy emphasizes the importance of bringing speech components into the audible range and restraining speech components below the discomfort level of the hearing aid users. In addition, although these conclusions were obtained using a hearing aid with two compression channels, it is conceivable that the same principles could be applied to hearing aids with more compression channels. Future studies are needed to verify these assumptions.

Summary and Conclusion

A two-channel experimental hearing aid was built to investigate the rankings of input–output functions for listeners with mild to moderately severe flat or sloping hearing loss when the adjustment of volume control is not possible or undesirable. The hearing aid hardware, level detectors, time constants, compression ratios, frequency responses, number of compression channels, instantaneous SNR between speech and babble, and the ability to recognize speech in multitalker babble within a listener group were kept constant.

The general findings of this study support previous findings that audibility is essential for understanding low-level speech and that compressors with low-compression ratios do not degrade or enhance speech recognition for conversational speech compared with linear amplification.
when the audibility of speech components is similar across conditions. The specific findings, which should apply to well-designed two-channel compression hearing aids with adaptive time constants and volume control not manipulated, are as follows:

1. WDRCs with low compression ratio compression regions extending to high input levels or WDRCs with high compression thresholds for the compression limiting regions are the most appropriate input-output functions for hearing aid users with flat and mildly sloping sensorineural hearing loss.

2. Curvilinear compression appeared to be the most appropriate for hearing aid users with flat hearing loss and very narrow dynamic ranges.

Acknowledgments

This study was conducted at Northwestern University in partial fulfillment of the doctoral degree requirements for the first author. We would like to thank Ettymotic Research for sponsoring the equipment and providing technical support to make this project possible and the American Academy of Audiology for the Student Investigator Grant. We would also like to thank Larry Revit for his guidance in making recordings of the speech testing materials and Greg Shaw and Dan Maps-Riordan for programming support. In addition, thanks go to Rachael Fischer and Tarez Graban for editorial help.

References


Received April 27, 2004
Revision received October 24, 2005
Accepted August 16, 2006
DOI: 10.1044/1092-4388(2007/022)

Contact author: King Chung, who is now with the Department of Speech, Language, and Hearing Sciences, Purdue University, Heavilon Hall B32, West Lafayette, IN 47907. E-mail: kingchung@purdue.edu.

Mead C. Killion is no longer affiliated with Rush University. Laurel A. Christensen is no longer affiliated with Etymotic Research. Her current affiliations include GN ReSound, Glenview, IL; Northwestern University; and Rush University.