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Behavioral assessment of adaptive feedback equalization in a digital hearing aid

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Abstract—An evaluation was made of the efficacy of a digital feedback equalization algorithm employed by the Central Institute for the Deaf Wearable Adaptive Digital Hearing Aid. Three questions were addressed: 1) Does acoustic feedback limit gain adjustments made by hearing aid users? 2) Does feedback equalization permit users with hearing-impairment to select more gain without feedback? and, 3) If more gain is used when feedback equalization is active, does word identification performance improve? Nine subjects with hearing impairment participated in the study. Results suggest that listeners with hearing impairment are indeed limited by acoustic feedback when listening to soft speech (55 dB A) in quiet. The average listener used an additional 4 dB gain when feedback equalization was active. This additional gain resulted in an average 10 rationalized arcsine units (RAU) improvement in word identification score.

Key words: *acoustic feedback, digital feedback equalization algorithm, evaluation, wearable adaptive digital hearing aid.*

INTRODUCTION

The objective of this study was to evaluate the efficacy of a feedback equalization (FBE) algorithm implemented on the Central Institute for the Deaf's (CID) Wearable Adaptive Digital Hearing Aid (WADHA).

Acoustic feedback is a familiar problem to those working with people who use hearing aids. Using a behind-the-ear (BTE) aid with a nonvented, well-fit earmold, Dyrlund and Lundh (1) reported that an average gain reduction of about 10 dB from prescribed gain was required to eliminate acoustical feedback. The limitations were greatest for frequencies above 1,000 Hz. Dyrlund (2) reported a limit on the hearing loss that could be managed without acoustical feedback for children with profound hearing impairment using BTE aids and nonvented acrylic earmolds. His guidelines for maximum hearing loss manageable before feedback show that for 1 kHz, maximum loss is 100 dB HL, and the upper intensity limit decreases as frequency increases. Grover and Martin (3) reported gains of about 50 dB before feedback oscillation using nonvented, acrylic earmolds and a BTE configuration. They compared the sound pressure level at the hearing aid microphone with that of a probe tube inserted into the ear canal. While many hearing losses can be accommodated with this amount of gain, the increased incidence of steeply sloping audiometric configurations gives rise to greater need for open or vented fittings, to provide adequate high frequency amplification without over-amplifying the low frequency region where hearing may be normal or near-normal.

The most common remedy for acoustical feedback is to simply reduce the gain of the hearing aid, which in turn reduces the intensity of the sound that may feedback to the microphone. This method is often employed by hearing instrument wearers. That is, many hearing aid wearers will adjust the volume control of their aids by turning the gain up until

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audible feedback occurs, and then turn the gain down until feedback ceases. Although use of this method of gain adjustment is widespread, and even recommended by some dispensers, it is a clear indication that the gain that hearing instrument wearers use is limited by the acoustic feedback threshold. In addition, Cox (4) advised against using the threshold of feedback as the user gain level because of the effects on the frequency response of the hearing aid output. She noted that by setting the gain control to a position just below that which would cause audible feedback, there occurs suboscillatory feedback that results in the formation of erratic peaks in the frequency response of the hearing aid. Skinner (5) suggested that insertion gains should be 4–8 dB less than values at which audible feedback occurs to avoid the deleterious effects of suboscillatory feedback. Yanick (6) noted that hearing aid wearers who adjust hearing aid gain to the threshold of audible feedback may be receiving distorted speech (both spectrally and temporally) as a result of transient distortion. Preves (7), discussing this phenomenon, stated that when formant transitions are near the frequencies of the resonant response peaks of a hearing aid operated just below acoustical feedback oscillation, they may become severely distorted and detract from their perception by listeners with hearing impairment.

Consequently, an adaptive feedback equalization algorithm that suppresses acoustic feedback while maintaining an appropriate target gain function may provide significant benefit to many hearing aid wearers. Hearing aid users may then select gain levels that amplify signals to comfortable levels without feedback. Further, adaptive feedback equalization may smooth the erratic peaks that occur when listeners select volume control levels that result in suboscillatory feedback. The smoothing of response may also improve speech identification performance. In the present study, three questions were addressed: 1) Does acoustic feedback limit gain adjustments made by hearing aid users? 2) Does feedback equalization permit users with hearing impairment to select more gain without feedback? and, 3) If more gain is used when feedback equalization is active, does word identification performance improve?

METHOD

Subjects

Nine subjects with hearing impairment (5 male and 4 female) having a mean age of 63.4 years (range 39–76 years) participated in this study. **Table 1** lists the pure-tone thresholds of the subjects. All nine subjects were experienced hearing aid wearers—five BTE and four in-the-ear (ITE). All subjects had sensorineural hearing losses, except Subject #1 who had a mixed hearing loss.

Wearable Adaptive Digital Hearing Aid

Figure 1 is a schematic of the digital hearing aid (DHA) used in this experiment. The DHA is a four-channel hearing aid. Each channel is specified by the first "Filter" in the Filter-Limit-Filter configuration. Gain shape and channel limit are specified in the "Limit" box. The final "Filter," shaped identically to the first filter, rejects any in-band harmonic distortion generated by the limiting process. Limiting in this evaluation was achieved using peak-clipping. The Noise Reduction feature was not active in this sequence of experiments. Details of the adaptive feedback equalization algorithm are described in a companion paper in this issue (8). In principle, the digital FBE algorithm suppresses feedback by adaptively canceling the feedback path. The process is illustrated schematically in **Figure 2**. The acoustical feedback pathway (H) represents sound energy that has leaked from a vent or around an earmold or hearing aid. The sound escapes from the ear canal (Y), and travels back to the hearing aid microphone (M). The input signal to the amplifier circuitry (G) is monitored at point (A). The output signal from (G) is monitored at point (B). A 1.8 ms delay is introduced at (G). Signal components at (B) that are correlated with those at (A) are considered to be derived from the feedback path (H). The coefficients of the FBE filter are subsequently adapted to reduce the correlation to zero. The FBE will adapt to the feedback signals whether the hearing aid starts out in a state of oscillation or begins to oscillate after the aid has been on for some time. The adaptive process initially may take 1–1.5 seconds to identify the feedback signals and adapt to them; however, any changes that occur in the

Table 1.
Pure-tone air conduction thresholds (dB HL re: ANSI, 1969).

Subject #	Age Years	Sex	Ear	Frequency, Hz 250	500	1000	2000	4000	8000
1	57	F	R	80	75	75	80	80	95
			*L	60	70	65	70	70	90
2	66	M	R	15	10	55	55	60	70
			*L	15	15	60	60	55	55
3	69	M	R	20	15	50	80	90	85
			*L	15	15	45	60	70	100
4	68	F	R	60	65	75	65	65	105
			*L	40	60	80	65	65	80
5	76	M	R	10	10	20	40	65	90
			*L	10	20	25	50	65	90
6	39	F	R	40	60	80	80	75	105
			*L	35	50	75	75	75	95
7	65	M	R	30	45	55	55	55	65
			*L	35	40	60	55	55	55
8	71	F	R	35	50	55	75	95	NR
			*L	30	40	50	65	85	90
9	60	M	*R	45	55	70	70	70	100
			L	30	50	70	75	65	100

*ear tested

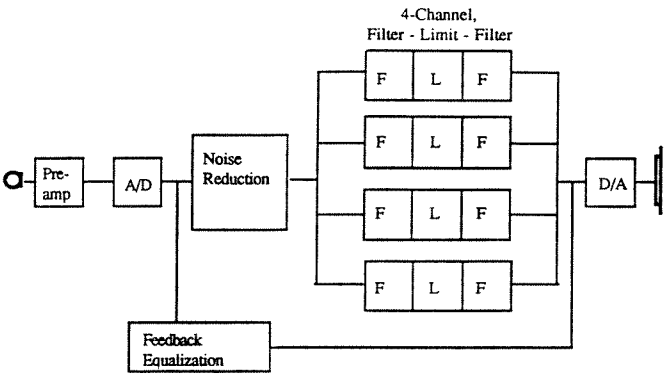

Figure 1.
Schematic illustration of the Digital Hearing Aid.

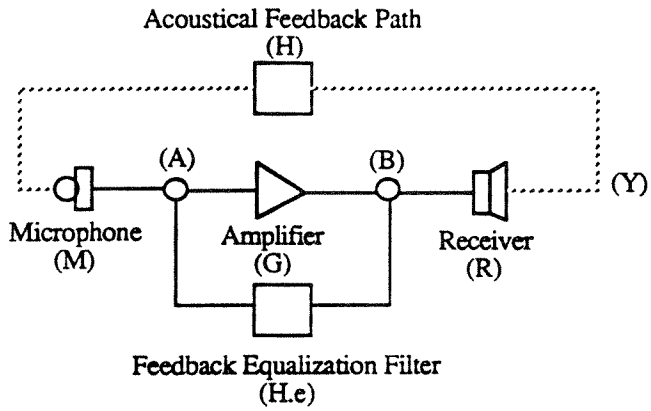

Figure 2.
Schematic illustration of feedback path and feedback equalization filter.

feedback pathway are adaptively canceled at a much faster rate (approximately 100 msec).

Test Room and Equipment

All listening tasks were carried out in a sound-treated room. Stimuli were presented from a loudspeaker located 1 m in front of the subject at 0° azimuth. Signal levels were calibrated in dB A (slow) at the position corresponding to that of the center of the head of the listener if the listener were present. Real-ear probe-tube measurements were made at 45° azimuth using a second loudspeaker, also 1 m from

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the subject. The real-ear measurement system consisted of a reference microphone clipped to the earlobe of the subject, and a soft cilastin probe-tube placed alongside the earmold in the external ear canal, within 5 mm of the tympanic membrane. Insertion depth of the probe tube was determined using the acoustic method as described by Gerling and Engman (9), which involves probe-tube insertion to a depth of 10 mm beyond the 6 kHz standing wave null. The subjects performed listening tasks aided monaurally, with the contralateral ear occluded with an EAR Noise Filter™ earplug (Cabot Corp., Indianapolis, IN).

Speech Stimuli

Subjects adjusted hearing aid gain control settings while listening to excerpts from the Connected Speech Test (CST) (10), which utilized a female talker. Word identification scores were determined using Pascoe High Frequency Word Lists (PHFWL), which utilized a male talker. Each PHFWL consists of 50 monosyllabic words containing a large proportion of high frequency consonants. Each word was presented with the carrier phrase, "Please write . . . .," with a 4-sec gap between presented words. Subjects wrote their responses on answer sheets, which were scored at a later time. In competing noise conditions, multitalker babble was presented at a 6 dB signal-to-babble ratio (SBR).

Determination of Gain Control Setting

Subjects adjusted the gain control of their own hearing aids (Part I) and a simulation of their own aids using the CID WADHA (Part II) in quiet and multitalker babble while listening to soft speech signals (55 dB A). Subjects were instructed to adjust the hearing aid gain control so that speech was perceived to be maximally intelligible.

In Part I, subjects were monaurally aided with one of their own hearing aids and earmolds. The subjects were instructed to adjust the gain control of their own hearing aid such that soft speech signal (CST paragraphs presented at 55 dB A) was perceived as most intelligible. Once the gain selection was made, the real-ear insertion response was obtained for input levels of 55, 70, and 85 dB sound pressure level (SPL). This procedure was performed for two listening conditions: speech-in-quiet and speech-in-babble. The order of experimental conditions was counter-balanced across subjects. The initial condition was replicated as a retest measure.

In Part II, the WADHA was configured to simulate the subject's own hearing aid by using target gain and saturation output values based on the real-ear measurements made in Part I. The real-ear insertion responses corresponding to the 55 dB A input composite noise signal were used to derive target gain values to be programmed into the WADHA. The real-ear insertion responses corresponding to the 85 dB A input composite noise signal supplied the values that were used to program the maximum power output of the WADHA configurations. The four WADHA memories were configured as follows:

Memory A: Simulation of subject's own aid in *quiet,* FBE **Off**.

Memory B: Simulation of subject's own aid in *quiet,* FBE **On**.

Memory C: Simulation of subject's own aid in *babble,* FBE **Off**.

Memory D: Simulation of subject's own aid in *noise,* FBE **On**.

The WADHA gain control was active during all listening tasks and measurements. The gain control has 16 steps of 2 dB. The target gain settings are achieved when the volume control is set at position "7." Once the WADHA was configured as above, the subjects made gain control adjustments for each of the four memories as before, and real-ear probe-tube measurements were made at the selected gain control settings using 55, 70, and 85 dB SPL composite noise input levels. The selected gain control settings were recorded for each memory. It should be noted that BTE hearing aid wearers retained their own earmolds for the WADHA evaluations while ITE hearing aid wearers were provided with custom silicone-shell earmolds. Fit measures obtained at 50 Hz intervals resulted in WADHA fit accuracy of 4.9 dB root mean square (rms) between 250 Hz and 6,000 Hz and 4.3 dB between 500 Hz and 2,500 Hz.

Word Identification Assessment

Percentage correct word identification was assessed using the PHFWL at a signal presentation level of 55 dB A. The listening conditions were varied randomly such that some of the trials required listening in quiet and others in a background of multitalker babble noise (+6 dB SBR).

The hearing aid (either the subject's own or WADHA simulation) was adjusted to the volume control setting that was selected previously, for each listening condition. The subjects wrote their responses on answer sheets, which were scored at a later time.

RESULTS

Recall that the first objective was to determine whether acoustic feedback limited gain adjustments made by the subjects when listening to soft speech in quiet and soft speech in babble noise at 6 dB SBR. In Part I of this study, all subjects responded similarly to the initial instructions to adjust their own hearing aid such that the speech stimulus is perceived as maximally intelligible. All subjects turned up their hearing aid until audible feedback was produced and then turned down the gain somewhat. When subjects were fitted with the WADHA simulation of their own aid with FBE Off, similar adjustments of the gain control were observed. Thus, it appeared that acoustic feedback was limiting the usable gain range of all subjects in this study.

The second objective of this study was to determine whether FBE permits users to access additional gain when listening to soft speech in quiet and babble noise. Two comparisons addressed this issue. First, comparisons were made between FBE On and FBE Off WADHA conditions. This comparison provided data regarding the efficacy of FBE within the prototype, fully digital amplification condition. Second, comparisons were made between the WADHA FBE On and Own-Aid conditions. These latter comparisons provided guidelines regarding the magnitude of gain change listeners may need within existing conventional amplification.

Gain Changes (FBE On–FBE Off) Implemented Using WADHA. **Figure 3** and **Figure 4** illustrate the amount of additional gain subjects used in the FBE On condition compared with the FBE Off condition when listening to soft speech in quiet and soft speech in babble, respectively. The box plots illustrate the median (waist of box plot), interquartile range (ends of box), 10th and 90th percentile range (whiskers), and outlier points. In quiet, the median difference in gain between the FBE On and FBE Off conditions ranged from 0 dB at 6,000 Hz to 5 dB at 500 Hz. Over the major speech frequencies, the median gain adjustment was approximately 4 dB. In babble noise, the median difference in gain between the FBE On and FBE Off conditions ranged from −1 dB at 6,000 Hz to 4 dB at 1,000 Hz. Over the major speech frequencies, the median gain adjustment was varied between 0 dB and 2 dB. An example of the effects of FBE on the electroacoustic output of WADHA is illustrated in **Figure 5**. The WADHA real-ear frequency response in the FBE On

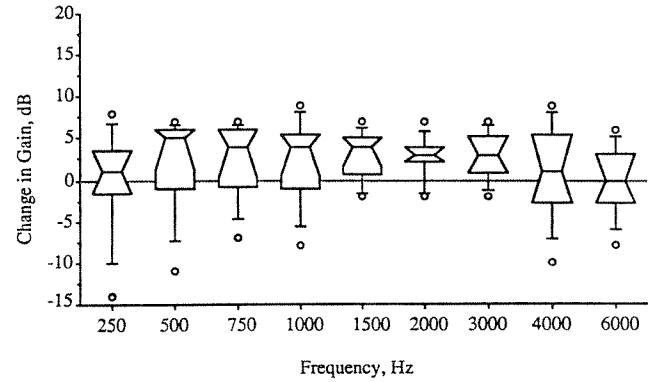

Figure 3.
Box plot distribution of averaged difference in insertion gain between the WADHA FBE On and FBE Off conditions, selected by subjects listening to soft speech in quiet as a function of frequency. The median gain difference (FBE On–FBE Off) is represented by the waist of the box plot. The box extends beyond the median to the interquartile values. The straight-line "whiskers" extend to the 10th and 90th percentile values. Outliers (≤ 10th or ≥ 90th percentile) are represented by circles.

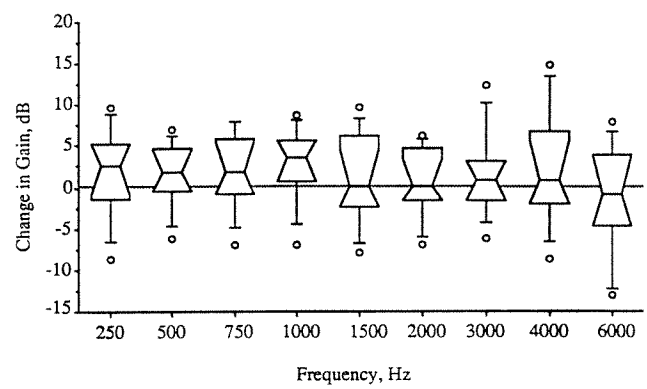

Figure 4.
Box plot distribution of averaged difference in insertion gain between the WADHA FBE On and FBE Off conditions, selected by subjects listening to soft speech in babble as a function of frequency.

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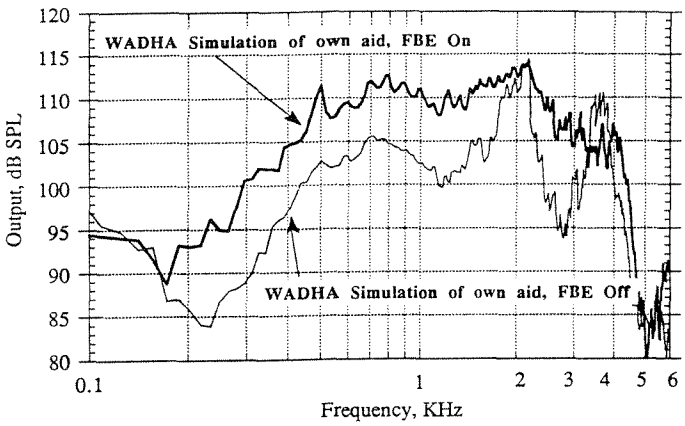

Figure 5.
Frequency response curves for WADHA simulations of own aid in FBE Off and FBE On conditions (Subject #1).

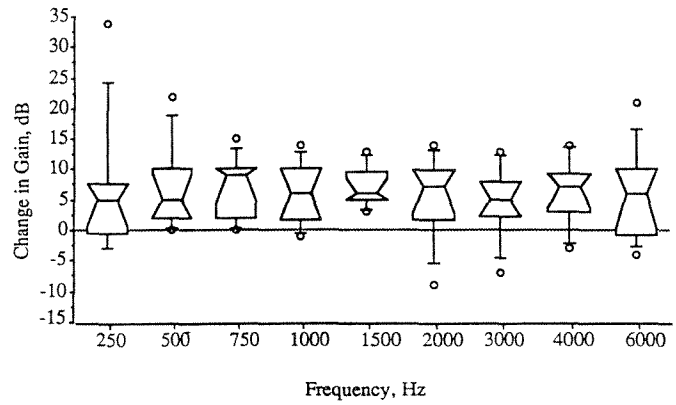

Figure 6.
Box plot distribution of averaged difference in insertion gain between the WADHA FBE On and own aid conditions, selected by subjects listening to soft speech in quiet as a function of frequency.

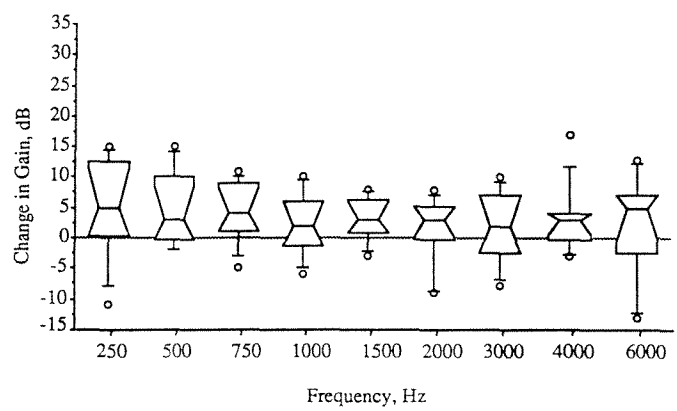

Figure 7.
Box plot distribution of averaged difference in insertion gain between the WADHA FBE On and own aid conditions, selected by subjects listening to soft speech in babble as a function of frequency.

and FBE Off conditions are shown for Subject 1. When listening to connected speech in quiet in the FBE Off condition, Subject 1 selected a WADHA volume control setting that produced audible feedback. The frequency response demonstrates large resonant peaks centered around 2,000 Hz and 3,500 Hz, and marked minima between 1,000 Hz and 2,000 Hz and close to 3,000 Hz. When FBE is activated, the resonant peaks are smoothed, and up to 7 dB of additional gain is available without audible acoustic feedback.

Gain Changes (FBE On-Own Aid) Implemented Using the WADHA Compared with Own Aid. **Figure 6** and **Figure 7** illustrate the amount of additional gain subjects used in the FBE On condition compared with their own aid when listening to soft speech in quiet and soft speech in babble, respectively. In quiet, the median difference in gain between the FBE On and FBE Off conditions ranged from 5 dB at 250 Hz, 500 Hz, and 3,000 Hz to 9 dB at 750 Hz. Over the major speech frequencies, the median gain adjustment was approximately 6 dB. In babble noise, the median difference in gain between the FBE On and FBE Off conditions ranged from 2 dB at 1,000 Hz and 3,000 Hz to 5 dB at 250 Hz and 6,000 Hz. Over the major speech frequencies, the median gain adjustment was varied between 0 dB and 3 dB.

The third objective of this study was to determine whether the additional gain used in the FBE On condition results in improved word identification performance when listening to soft speech in quiet and babble noise conditions. **Table 2** provides the word identification performance of the subjects when listening with their own hearing aid or a digital simulation of their own aid with the WADHA in quiet and +6 dB SBR. The scores reported have been transformed from percentage correct to rationalized arcsine units (RAU) to normalize the variability in the data (11). Examination of the group means reveals that subjects performed better in quiet than in babble (using either their own aid or a WADHA simulation of their own aid). Subjects also performed better in the FBE On condition compared with the FBE Off condition in quiet and in babble. **Figure 8** illustrates the linear regression of the word

Table 2.
Word Identification Performance (RAU).

	Subject's Own Aid		Digital Simulation of Own Aid			
	Quiet	Noise*	Quiet		Noise*	
Subject #			FBE On	FBE Off	FBE On	FBE Off
1	72.80	89.10	91.90	77.00	89.10	66.80
2	62.90	53.60	81.50	68.70	59.20	57.30
3	53.60	51.80	68.70	64.80	53.60	48.20
4	77.00	66.80	83.90	79.20	81.50	68.70
5	89.10	77.00	107.20	98.40	77.00	70.70
6	53.60	50.00	59.20	57.30	44.50	46.40
7	68.70	59.20	83.90	74.90	70.70	77.00
8	53.60	42.70	91.90	72.80	51.80	55.50
9	53.60	55.50	64.80	48.20	57.30	55.50
Average	65.00	60.60	81.40	71.30	65.00	60.60

*Noise (babble) = +6 dB SBR

identification scores of the subjects using the WADHA simulation of subjects' own aid frequency response on the word identification scores obtained by subjects using their own hearing aid in the quiet listening condition. A small (mean difference = 6.27 RAU, $t = 2.785$, $p = 0.019$) but significant improvement in word identification score is observed when subjects' own aid scores are compared with the WADHA simulation FBE Off condition. This result is not entirely unexpected, as the majority of subjects were able to use additional gain without feedback under the WADHA simulation (FBE Off) compared with that available with their own aid. This was particularly noticeable in four subjects who used ITE hearing aids. An additional significant improvement in word identification performance is observed when the FBE On and FBE Off conditions are compared. A directional t-test for correlated samples was performed on the data, and revealed the group performance increase of 10.2 RAU observed in the FBE On condition to be significant ($p = 0.0005$; $t = 5.066$).

Figure 9 illustrates the linear regression of the word identification scores of the subjects using the WADHA simulation of subjects' own aid frequency response on the word identification scores obtained by subjects using their own hearing aids in the babble listening condition. No significant differences were observed in group mean performance.

DISCUSSION

The results from this study suggest that acoustic feedback restricts the volume control adjustments that hearing aid wearers can make to compensate for loss of audibility in soft speech environments. The current feedback equalization algorithm has been shown to produce gain margins of 15–20 dB in the laboratory (8). However, it should be remembered that these values are obtained under ideal testing situations using a mannequin that does not move. With real subjects who breathe, talk and chew, practical gain margins are likely to be less than 15–20 dB. Examination of **Figure 6** and **Figure 7** suggests that gain margins of up to 10 dB might be required to accommodate the additional gain needs of 75 percent of the subjects used in this study. It should be noted that none of the subjects in this study used a vented earmold. Thus, the needed gain margins in vented mold situations may start to encroach on the gain margin limits of the adaptive algorithm. Gatehouse (12), for example, has shown that gain must be reduced by 15–20 dB when a 2 mm parallel vent is introduced into an earmold connected to a BTE hearing aid with a forward-facing microphone. Gatehouse also observed that the listening environment influences the amount of gain that is available without audible feedback. For example, a BTE hearing aid with a forward-facing

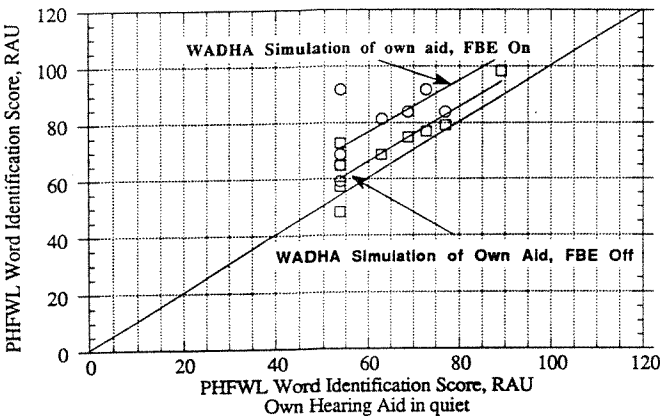

Figure 8.
Linear regression of PHFWL word identification scores (RAU) for WADHA simulations of own aid in quiet for FBE On and FBE Off conditions on PHFWL scores (RAU) for own aid in quiet.

$SCORE_{WADHA\ FBE\ On} = 22.027 + 0.91\ SCORE_{Own\ Aid},\ R^2 = 0.605$

$SCORE_{WADHA\ FBE\ Off} = 7.863 + 0.975\ SCORE_{Own\ Aid},\ R^2 = 0.776$

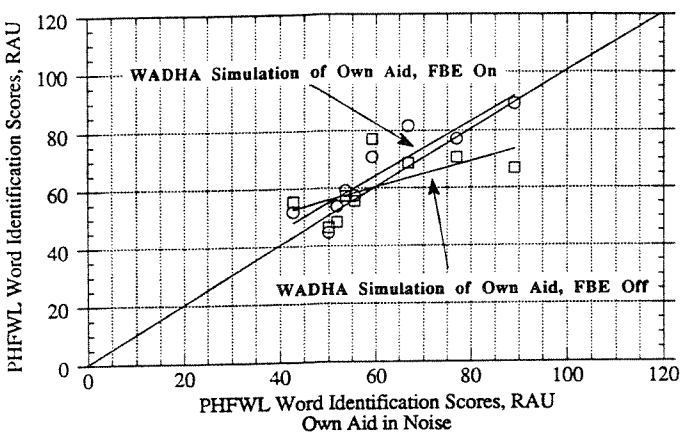

Figure 9.
Linear regression of PHFWL word identification scores (RAU) for WADHA simulations of own aid in babble for FBE On and FBE Off conditions on PHFWL scores (RAU) for own aid in babble.

$SCORE_{WADHA\ FBE\ On} = 7.71 + 0.94\ SCORE_{Own\ Aid},\ R^2 = 0.83$

$SCORE_{WADHA\ FBE\ Off} = 33.34 + 0.451\ SCORE_{Own\ Aid},\ R^2 = 0.393$

microphone could support 56.9 dB insertion gain without acoustic feedback in the presence of speech-shaped noise, 53.1 dB gain in the presence of a 250 Hz narrow-band noise, and 64.3 dB in the presence of a 2,000 Hz narrow-band noise. In this investigation, it was noted that subjects turned their own hearing aids up somewhat in the speech-in-babble condition. **Figure 10** illustrates the median gain

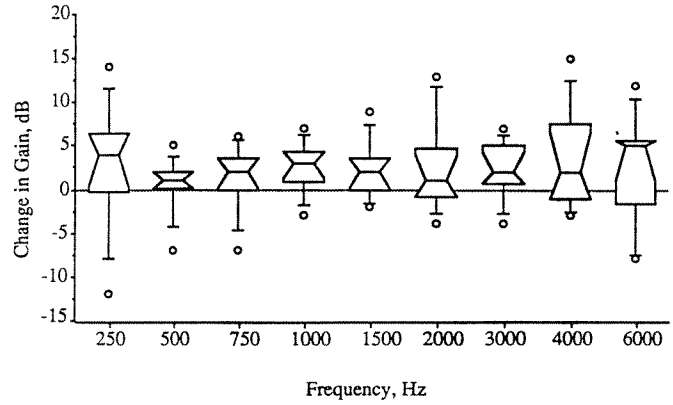

Figure 10.
Box plot distribution of averaged difference between insertion gain selected by subjects listening to soft speech in babble compared with soft speech in quiet as a function of frequency.

adjustments subjects made to their own hearing aids when listening to soft speech in quiet compared with listening to soft speech in babble. The gain adjustment resulted in a median increase in gain in babble of approximately 2 dB over the major speech frequencies. It is possible that the presence of the multitalker babble served to stabilize an otherwise unstable acoustic environment, thus permitting some additional gain in the speech in babble condition. Further work in this area is required.

The additional gain that subjects were able to achieve in the FBE On condition translated into significant improvements in word identification performance in quiet. The approximate 16 RAU improvement in mean score in the FBE On condition compared with the subjects' own aid condition suggests that if FBE were available in commercial aids, hearing aid wearers would derive perceptible improvement in word identification performance. The lack of significant improvement in word identification performance in babble is not surprising. The limiting factor in the babble condition was the SBR, which would have been largely unaffected by the linear gain changes demonstrated in this study.

In conclusion, subjects in this study were able to take advantage of additional gain without acoustic feedback provided by the FBE algorithm. The additional gain resulted in improved word identification performance in quiet. The gain adjustments required by these subjects appear to be within the capabilities of the technology.

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