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Evaluation of a feedback equalization algorithm in the CID Wearable Digital Hearing Aid

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Evaluation of a Feedback Equalization Algorithm in the CID Wearable Digital Hearing Aid.

An Independent Study Project

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INTRODUCTION

Advances in hearing instrument technology have resulted in a number of improvements that have benefitted and addressed the concerns of both hearing instrument wearers and dispensers. Miniaturization of circuitry has given birth to devices much reduced in size, yet possessing high output capabilities (although reduced headroom in many power ITC instruments), improved control of the frequency response, output limiting strategies to control for user discomfort levels, etc. (Pollack, 1988). In addition, the advent of micro-chips, and their subsequent application to hearing aids has allowed for greater innovation via signal processing. Improved speech intelligibility, reduction of background noise, and overall sound quality improvements are just a few of the changes possible with signal processing techniques (Levitt, 1991).

However, despite these innovations, certain problems that have existed for years persist and continue to represent limitations in achieving high levels of user satisfaction with hearing instruments. Among the most common of these lingering problems is acoustical feedback, often referred to as "squealing", "whistling", or "howling", which occurs most often in cases of an open, vented, or poorly fitting earmold and/or hearing losses requiring high gain (Lybarger, 1982).

Acoustical feedback occurs when a portion of the amplified sound from a hearing instrument receiver leaks out of the ear canal via a vent or unintentionally around a poorly fitting aid or earmold and travels back to the input (microphone), where it is amplified over and over again until the circuit begins to oscillate. This oscillation results in the familiar squeal experienced by many hearing aid wearers (Preves, 1988). If the hearing aid has low-moderate gain, that sound which has leaked will probably lose much of its energy by the time it finds its way back to the microphone, and not result in an oscillation. However, with higher gain aids, the leaking sound reaches the microphone at a greater intensity and the sound may be amplified over and over. As the cyclic process of re-amplification occurs, the resonant response peaks of the hearing aid will be the components of greatest intensity in the feedback signal. When these components are in-phase with incoming sounds at the hearing aid microphone, they may add together resulting in a signal of even greater intensity transmitted to the hearing aid circuitry. When the intensity becomes great enough, the circuit of the hearing aid becomes unstable, begins to oscillate, and provides a less than optimal signal for the wearer.

There are numerous reports documenting the limitations imposed on accessible gain as a result of acoustical feedback. Several investigators have examined gain limitations for hearing aids worn by profoundly hearing-impaired persons. Using a BTE aid, with a non-vented, well-fit earmold, Dyrlund and Lundh (1989) reported that acoustical feedback required an average gain reduction of about 10 dB from prescribed gain. The limitations were greatest for frequencies above 1000 Hz. Dyrlund , (1989) reported a limit on the hearing loss that could be managed without acoustical feedback for profoundly hearing-impaired children using BTE aids and non-vented 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 limit decreases as frequency increases. This suggests that fitting a sloping severe to profound hearing loss would be quite difficult without some problems with acoustical feedback.

Grover and Martin (1974) reported gains of about 50 dB before feedback oscillation using non-vented, acrylic earmolds and a BTE configuration. They compared the sound pressure level at the hearing aid microphone to 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. In addition, moderate to severe hearing losses are increasingly being fit with ITE or canal instruments whose susceptibility to acoustical feedback will be discussed later in this paper.

Gatehouse (1989) reported reductions of 15-20 dB in the usable gain of a BTE aid with a forward facing microphone and a 2.0 mm parallel vent, when compared to the non-vented condition. Even with a reduction in vent diameter (0.8 mm), significant reductions (10 dB) in accessible gain were noted, contrary to some claims that a 0.8 mm vent has no acoustical effects.

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 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 hearing instrument wearers use is limited by the acoustic feedback threshold. In addition, Cox (1982) 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 which results in the formation of erratic peaks in the frequency response of the hearing aid. Skinner (1988) suggested that insertion gains should be 4 to 8 dB less than values at which audible feedback occurs to avoid the deleterious effects of suboscillatory feedback.

Yanick (1977) 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 (1988), 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 hearing-impaired listeners.

A number of methods have been proposed to control acoustical feedback. These have ranged from physical alterations of the aid or earmold to electronic methods utilizing complex signal processing paradigms. Since the feedback path extends from the output of the hearing aid receiver back to the microphone, increased physical separation of these two components should reduce the occurrence of feedback, by virtue of the inverse square law (i.e., reductions in intensity with increased distance from the sound source). The large separation between microphone and receiver output (up to 18 inches) in body-worn aids accounts for the large gains which can be achieved with little or no feedback. Conversely, in-the-ear (ITE) and canal instruments are most susceptible to feedback since there may be as little as 0.25 inches between the microphone and receiver output (Pollack, 1988).

Grover and Martin(1974) reported high levels of attenuation in the feedback path with the use of closed acrylic earmolds. Using either shell or skeleton type earmolds, they were able to achieve upwards of 50 dB or more gain in 90% of the molds studied. Tucker et al, (1981) found the use of soft earmold material allowed greater gain availability than hard acrylic earmolds.

In the routine fitting of hearing aids it is often desirable to utilize a vent to reduce the feeling of occlusion ("stuffed up" feeling due to occlusion of external ear canal), to attenuate the low frequency gain (the vent acts as a high pass filter), and to reduce ambient noise amplification. The vent, however, serves as a direct acoustic pathway from the output of the receiver to the hearing aid microphone, and can severely limit the usable gain due to acoustical feedback. Several investigators have explored reduced vent diameter or altered microphone placement to control acoustical feedback when a vent is desired. Gatehouse(1989) investigated the effects of microphone orientation and vent size in vented earmolds on acoustical feedback. Using aids with up to 64 dB gain, he found a forward facing microphone less susceptible than a downward facing one and that a smaller vent (0.8mm) can control feedback more than a larger (2mm) vent. Lybarger (1975) reported greater gain before the onset of acoustical feedback using a smaller diameter Pressure Venting Valve (PVV; #4 vs #1). He also found that having the microphone placed behind the pinna allowed for increased amounts of usable gain, rather than in front of the pinna. Curran (1992) reported improvements in feedback margin using a combination of reduced low frequency amplification and/or shortening and widening the size of a vent for canal instruments. The gain margin was 4-9 dB for 2 clients who had been unsatisfied with the amount of high frequency amplification obtainable before feedback.

Acoustical feedback can be regarded as a high frequency phenomenon, in part because higher frequency sounds have shorter wavelengths which allow for easier escape around a poorly fitting aid or earmold or through a vent. In addition, the resonant response peaks of most hearing aids occur at higher frequencies (1-4 kHz) (Pollack, 1988), and as discussed earlier, these peaks in the response are often the components of the feedback signal responsible for acoustical feedback, since they are of greatest intensity. Thus, one technique utilized for the control of acoustical feedback is to reduce the amount of high frequency amplification. This can be achieved in several ways, but the most common is through the use of low-pass filtering. Although the feedback problem may be controlled, the negative side effect is usually insufficient high frequency gain for the hearing aid wearer (Preves, 1988).

Another electronic method of feedback control is through the use of a notched filter at the frequency where feedback oscillation is occurring. The problems associated with using a notched filter are that: 1) feedback may be present at more than one frequency; and 2) a "hole" is often left in the frequency response where the filter has acted, often in a critical region of high frequency amplification (Preves, 1988).

Recently, Dyrlund and Bisgaard (1991) reported feedback margin improvements of about 10-16 dB using a prototype digital feedback suppression system interfaced with conventional analog BTE and ITE hearing aids. Their system essentially reduces the effect of the feedback signal by adding an almost identical signal (opposite in sign) to the input, which can eliminate some of the feedback signal.

Several authors have described other electronic or signal processing methods for controlling acoustical feedback in either hearing aids or public address (PA) systems (Preves, 1988; Egolf, 1982). It should be noted that most of the techniques have been developed for public address systems and have not proven easily adaptable for use in hearing aids. These methods include phase shifting or compensating, input signal frequency shifting or modulation, and a time delay circuit. None of these techniques have been incorporated into regular hearing aid production. Egolf (1982) states that more research is required before their application to hearing aids can take place, while Preves (1988) contends that the difficulty has been one of inadequate component miniaturization. Irrespective of the reasons why, hearing instrument wearers and dispensers are still faced with the issue of acoustical feedback, leaving them with only a few choices (i.e., reduced gain, elimination of vents, reduced high frequency amplification, etc.), none of which are desirable.

As was stated earlier, digital signal processing has opened up new avenues of exploration in attempting to solve some of the existing problems with hearing instruments. This paper describes the results of an investigation which examined the efficacy of a feedback equalization algorithm (Engebretson, O'Connell, and Gong, 1990) incorporated into the Central Institute for the Deaf (CID) Wearable Digital Hearing Aid (WADHA). [See Appendix A for descriptions of the WADHA and the Feedback Equalization (FBE)] Specifically we wished to know if the feedback equalization would allow for greater usable gains when subjects listened to soft speech signals, and if so, whether or not this would improve speech intelligibility scores when compared to the subject's own hearing aids or a digital simulation of their own hearing aids.

METHODS & MATERIALS

SUBJECTS

Nine hearing-impaired subjects (5 male and 4 female) having a mean age of 63.4 years (range 39-76 years) participated in this study. Table I lists the pure-tone thresholds of the subjects. All nine subjects are experienced hearing aid wearers (5 BTE and 4 ITE). One of the subjects (subject #1) has a mixed hearing loss. All other subjects have sensorineural hearing losses.

TABLE I. Pure-tone air-conduction thresholds (dB HL re: ANSI, 1969) of the subjects in this study.

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

a = Denotes ear used for testing

TEST ROOM AND EQUIPMENT

All listening tasks were carried out in a sound-treated room whose layout is presented in Figure 3. Stimuli were presented from a loudspeaker located 1 meter in front of the subject at 0 degrees azimuth. Signal levels were calibrated for the position corresponding to where the center of the listener's head would be, in dBA (slow), with the listener absent. Real-ear probe-tube measurements were made at 45 degrees azimuth using a second loudspeaker, also 1 meter from the subject. The real-ear measurement system consisted of a reference microphone clipped to the subject's earlobe, and a soft cilastin probe-tube placed alongside the earmold in the subject's external ear canal, within 5mm of the tympanic membrane. Insertion depth of the probe tube was determined using the acoustic method as described by Gerling & Engman (1991), which involves insertion beyond the standing wave of a 6 kHz tone presented in the sound field. The subjects performed listening tasks aided monaurally, with the contralateral ear occluded with an EAR Noise Filter™ earplug.

SPEECH STIMULI

Subjects adjusted hearing aid gain settings while listening to excerpts from the Connected Speech Test (CST) (Cox, 1989), which utilized a female talker. Speech intelligibility scores were determined using Pascoe High Frequency Word Lists (PHFWL), which utilized a male talker. 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 second gap between presented words. Subjects wrote their responses on answer sheets, which were scored at a later time. Competing noise consisting of multi-talker babble was used in some of the trials in both the preferred gain selections and speech intelligibility parts of the investigation.

DETERMINATION OF PREFERRED GAIN SELECTIONS

Subjects made preferred gain selections using their own hearing aids (Part I) and a simulation of their own aids using the CID Wearable Digital Hearing Aid (WADHA) (Part II) in quiet and noise while listening to soft speech signals (55 dBA). The multi-talker babble was presented at 49 dBA, thus providing a 6 dB speech to noise ratio.

In Part I, subjects were seated before a loudspeaker, wearing their own hearing aids (monaurally), listening to the CST either in quiet or in a background of noise. The subjects were instructed to adjust the volume control of their own hearing aid such that the soft speech signal was most intelligible. Once the gain (volume) selection was made, the subject was placed at 45 degrees azimuth to a second loudspeaker and a real-ear probe-tube measurement was made for input levels of 55, 70, and 85 dB SPL. This was done for listening in quiet and in noise, and then a re-test was performed of either the quiet or noise condition, chosen randomly. Thus, real-ear response curves were obtained showing the preferred real-ear gain of the subjects own aids for soft speech in both quiet and noise, at various input levels.

In Part II, the WADHA was configured to simulate the subject's own hearing aid by using target gain values based on the real-ear measurements made in Part I. The frequency response curve corresponding to 55 dBA input for quiet and noise was used to derive target gain values to be programmed into the WADHA.

The frequency response corresponding to the 85 dBA input supplied the values that were used to program the maximum power output (MPO) of the WADHA configurations. The 4 memories of the WADHA were programmed with certain gain values for quiet and another set of gain values for noise, with the Feedback Equalization (FBE) active on 2 of the 4 memories.

The four memories of the WADHA 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 *noise*, FBE **OFF.**

Memory D) Simulation of subject's own aid in *noise*, FBE **ON.**

The volume control of the WADHA was active during all listening tasks and measurements. The volume control is a dial which can be rotated 16 points along the wheel. Each gradation point corresponds to a 2 dB change in output. Once the WADHA was configured as above, the subjects made volume control selections for each of the 4 memories as before, and real-ear probe-tube measurements were made at the selected volume control settings using 55, 70, and 85 dB SPL input levels. These selected volume control settings were recorded for each memory. The preferred gain selections made in Parts I and II were those used for determination of speech intelligibility scores.

ASSESSMENT OF SPEECH INTELLIGIBILITY

Speech intelligibility was assessed using the Pascoe High Frequency Word Lists (PHFWL) at a level of 55 dBA. The listening conditions were varied randomly such that some of the trials required listening in quiet and others in a background of multi-talker babble noise (6 dB S/N Ratio). The hearing aid (either 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

Table II summarizes the performance of the subjects when listening with their own hearing aid or a digital simulation of their own aid with the WADHA. The scores reported have been transformed from percentages to Rationalized Arcsine Units (RAU's; Studebaker, 1985) to normalize the variability in the data. Also listed are the values corresponding to changes in gain made by the subjects when the Feedback Equalization (FBE) was active. The volume control adjustment changes reflect volume control changes made by the subjects when the FBE was active, and the achieved gain changes reflect real-ear measurements of gain change with FBE active, averaged over 250 to 8000 Hz at octave and inter-octave frequencies. Examination of the group means reveals that subjects performed better in quiet (using either their own aid or a WADHA simulation of their own aid), and when FBE was active (in either quiet or noise).

Table II. Word Intelligibility Scores for subjects for listening with their own hearing aid or a digital simulation of their own aid. Scores reported are in Rationalized Arcsine Units (RAU's), converted from percentage correct scores.

	Subject's Own Hearing Aid		WADHA Simulation of Subject's Own Aid							
Subject	Quiet	Noise	Quiet FBE ON	Quiet FBE OFF	Gain Change ΔVC	Gain Change ΔRE	Noise FBE ON	Noise FBE OFF	Gain Change ΔVC	Gain Change ΔRE
1	72.8	89.1	91.9	77.0	+6 dB	+5.1 dB	89.1	66.8	+2 dB	+5.6 dB
2	62.9	53.6	81.5	68.7	+4 dB	+3.7 dB	59.2	57.3	0 dB	+0.8 dB
3	53.6	51.8	68.7	64.8	+4 dB	+3.2 dB	53.6	48.2	+8 dB	+7.3 dB
4	77.0	66.8	83.9	79.2	+4 dB	+2.5 dB	81.5	68.7	-2 dB	-7 dB
5	89.1	77.0	107.2	98.4	-2 dB	-1.3 dB	77.0	70.7	0 dB	+0.5 dB
6	53.6	50.0	59.2	57.3	+4 dB	+4 dB	44.5	46.4	+2 dB	+3.2 dB
7	68.7	59.2	83.9	74.9	+6 dB	+4.3 dB	70.7	77.0	-2 dB	-3.1 dB
8	53.6	42.7	91.9	72.8	+8 dB	+8.6 dB	51.8	55.5	+2 dB	+1.8 dB
9	53.6	55.5	64.8	48.2	+4 dB	-3.1 dB	57.3	55.5	+8 dB	+7.9 dB
	X=65	**X=60.6**	**X=81.4**	**X=71.3**	**X=4.2 dB**	**X=3 dB**	**X=65**	**X=60.6**	**X=2 dB**	**X=1.9 dB**

Δ VC = Volume control change made by subjects going from FBE OFF to FBE ON conditions.

Δ RE = Real-ear gain change achieved by subjects via volume control change going from FBE OFF to FBE ON conditions.

The first issue addressed is whether or not subjects prefer greater amounts of gain before feedback when listening to soft speech than is provided by their own hearing aids. Figure 4 illustrates both the desired and achieved gain changes between the FBE off and FBE on listening conditions in quiet. All but one subject (#5) desired greater gain with FBE active, ranging from -2 dB to +8 dB, as indicated by volume control adjustments. In addition, every subject (except #9) that desired greater gain realized that gain increase (within 1-2 dB) as measured in the ear canal. Figure 5 shows the same comparison as Figure 4, for the noise condition. Although the pattern is less obvious, 7 of 9 subjects either kept the same gain setting or increased the gain with the FBE active.

The second question addressed was whether or not the increased gain accrued from feedback equalization would result in improvements in the subjects' speech intelligibility scores for the Pascoe High Frequency Word Lists (PHFWL). Figures 6 and 7 show graphically the scores listed in Table II. Figure 6 represents the PHFWL scores for listening in quiet, comparing the FBE ON versus FBE OFF conditions. ALL subjects demonstrated an improvement in PHFWL scores when the FBE was active (range 2-19 RAU's), compared to the FBE OFF condition. A directional t-test for correlated samples was performed on the data, and revealed the group performance changes observed with FBE active to be significant (p=.01; t=5.14).

Figure 7 illustrates the scores for the noise condition. There is no clear pattern when comparing FBE OFF versus FBE ON scores, although 6 of the 9 subjects did demonstrate some improvement with the FBE active. A directional t-test for correlated samples revealed the group changes to be insignificant (t=1.44; p=.01).

Figures 8 and 9 compare the subjects' performance when listening with their own hearing aids to the WADHA simulation of their own aid. Since the subjects own aids do not have a feedback suppression mechanism, the performance with the WADHA is with the FBE OFF. In the quiet condition, all subjects but one (#9) demonstrated an improvement in PHFWL scores when listening with the WADHA. In noise, the subjects were roughly evenly divided, with 4 showing better performance with their own aids, 1 subject performing equally well with both, and 4 showing improvement with the WADHA.

Figure 10 illustrates the PHFWL scores for the subjects in all conditions. The highest score attained by each of the subjects in any condition was listening with a WADHA simulation of their own aid in quiet, with FBE active. The next highest score attained by the subjects was also while listening with the Wearable Digital Hearing Aid (WADHA), although the condition varied.

Figure 11 a, b, and c, represent correlations between conditions in the subjects' PHFWL scores. Fig. 11a illustrates the correlation between performance in quiet with the FBE ON versus FBE OFF with the WADHA simulation of their own aid. The correlation is reasonably high, with approximately 84% of the performance improvement explained by the activation of feedback equalization. Figures 11b and 11c show the correlation in subject's performance between listening with their own hearing aids and the WADHA simulation. As is clear, the variability in 11b is large, and there is no discernable correlation. There is some correlation seen in 11c (0.78) in quiet, when comparing subjects' own aid to WADHA performance, but is probably due to a single extreme score.

As was mentioned earlier, the CID Wearable Digital Hearing Aid (WADHA) allows the programming of any target gain values desired for octave and inter-octave frequencies from 250 to 8000 Hz. The digital simulation of the subjects' own hearing aids utilized target gain values derived from real-ear measurements made with the subjects wearing their own aids at preferred volume control settings. In order to assess the accuracy of the digital simulation (or of any target gain values programmed) the WADHA has an algorithm that measures fit accuracy by comparing the programmed gain against the measured real-ear gain, and calculating the difference. The comparisons are made every 50 Hz, from either 250 to 6000 Hz, or 500 to 2500 Hz, and calculated as a Root-Mean-Square (RMS) error value.

Table III lists the fit accuracy of the digital simulation for each subject, in either the quiet or noise listening condition. Fit accuracy ranged from an error of 1.3 dB to 10 dB, with a mean RMS error of 3.75 dB in quiet, and 4.5 dB in noise, across the bandwidth 500-2500 Hz. This means that the WADHA was able to achieve an accurate simulation of the subject's own hearing aid, within 5 dB or less, from 500 to 2500 Hz, and within 5.5 dB or less across the broadband 250-6000 Hz. Figures 12 and 13 illustrate the fit accuracy found in Table

III, measured across the two bandwidths, for listening in quiet and noise, respectively.

DISCUSSION

This study set out to examine if the Feedback Equalization (FBE) algorithm of the CID Wearable Digital Hearing Aid (WADHA) would allow greater accessible gain without feedback, and if so, whether or not those gain increases would result in improvement in speech intelligibility as measured using the Pascoe High Frequency Word Lists (PHFWL).

The initial listening condition involved the subjects utilizing their own hearing aids, with the volume control adjusted for maximum perceived intelligibility. It is interesting to note that virtually all subjects in this study adjusted their own hearing aids by turning the volume control up to the threshold of audible feedback, then decreased the gain slightly until the feedback ceased. The problems that may accompany a user gain setting at the threshold of feedback are discussed in the introduction of this paper. However, it would seem to be a clear indication that these hearing aid wearers desire greater gain (before audible feedback) than they are able to access with their own hearing aids, when listening to soft speech signals. All the subjects are experienced hearing aid wearers, and report being reasonably satisfied with the aids they currently wear, yet when provided the opportunity to access greater gain (i.e., with FBE ON), all subjects except one did so, at least when listening in quiet (mean increase= 3-4 dB; range= -2 to 8 dB). Examination of performance associated with the gain increases reveals a mean improvement of about 10 RAU's, which is approximately equivalent to a 10% increase, and represents a significant improvement.

The gain increase realized with the FBE active was not large (i.e., 3-5 dB), however, one could argue that by suppression of feedback at this gain setting, not only was audible feedback controlled, but perhaps some of the deleterious effects of suboscillatory feedback as well. It may be that the subjects actually desire only small increases in gain when speech signals are at a soft level, but when attempting to access this slight increase without feedback equalization, both audible and suboscillatory feedback may play a detrimental role in speech perception.

It should be noted that when discussing the RMS error of fit accuracy in the WADHA simulations, certain points need be understood. First of all, when target gain values are programmed into the WADHA, they are done so with a default volume control setting assumption of #7, which is the mid-point value on the volume control dial. Thus, the subjects have the option of turning the volume control dial up or down from the target gain by an equal amount (about 14 dB). The target gain values programmed into the WADHA were taken from real-ear measurements of the subjects' own aid output, at their preferred volume setting (see Table IV). Since virtually every subject turned his/her own aid to the maximum setting (before audible feedback), it follows then that gain values corresponding to the subject's own aid volume set at or near maximum were downloaded to a mid-point volume control setting on the WADHA. If the gain requirements of the subjects were large, then even small increases in the WADHA volume control may have produced an output too great for the receiver to transduce without distortion, and this may have contributed to the bigger gain errors seen with some of the subjects. Subjects 1, 4, 7, 8, and 9, were the only subjects to have gain errors that exceeded 5 dB across the bandwidth 500-2500 Hz, and subjects 1, 4, 6, and 9, were the only subjects to have gain requirements that met or exceeded 35 dB. This means that subjects 1, 4, and 9, all had large gain requirements and large gain errors. It is possible that the large gain requirements may have contributed to the large gain errors for these subjects. Other factors may have contributed as well, such as probe tube movement in the ear canal, and cannot be ruled out.

The equivocal performance results when listening in noise with the WADHA simulation comparing FBE OFF versus FBE ON are probably a reflection of the difficulty most hearing-impaired persons experience when listening to speech in a background of noise. Many hearing aid wearers report that they either turn down or turn off their hearing aids when in a noisy environment. However, many of the subjects in this study (6 out of 9) either kept the gain setting the same or turned it up with the FBE active in noise. This may point to a need for further research into the factors that contribute to speech perception in noise (i.e., Is there some aspect of the processing in the WADHA or FBE that reduces the annoyance of background noise?).

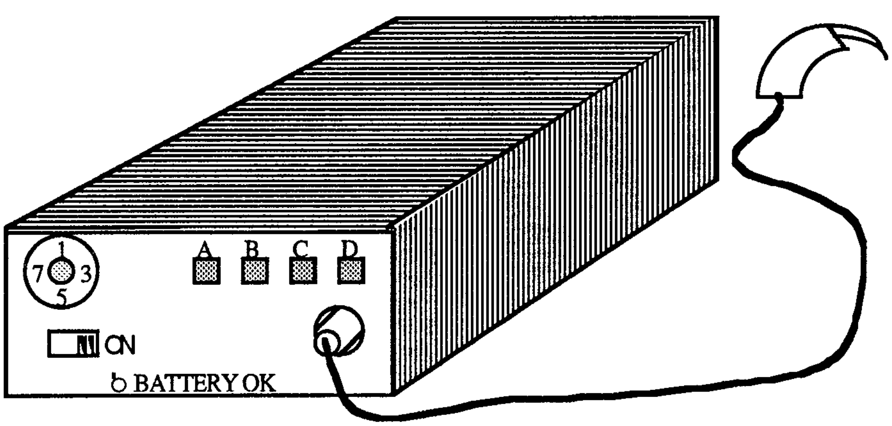
Clearly the subjects in this study demonstrated both increases in accessible gain and in speech perception scores with the Feedback Equalization (FBE) active, using the CID Wearable Digital Hearing Aid (WADHA), when listening to soft speech in quiet. Background noise complicates the issue somewhat, but small increases in gain were still attained by some of the subjects. The performance results in noise are equivocal, showing no significant change with the FBE active. Utilizing a larger number of subjects might have demonstrated the changes more clearly. However, this initial investigation is promising in the potential application of such technology to ear-level hearing instruments, for both greater accessible gain and improved speech perception when listening to soft speech signals.

APPENDIX

THE CID WEARABLE DIGITAL HEARING AID (WADHA)

The WADHA is a fully digital, four-channel prototype hearing aid that is programmed via hardwire connections using an IBM PS/2 personal computer interfaced with a custom signal generator and measurement system (PD-25). The hearing aid consists of a small box (about the size of a small cigar box), having the dimensions 16cm x 9cm x 4.5cm, and containing the digital circuitry (See Fig. 1). On one face of the unit is a volume control dial and 4 buttons that allow selection of any of the 4 memories in the hearing aid. The WADHA is hardwired to a conventional BTE hearing aid shell (3M Memory Mate™) which serves as the ear-level input and output transducers and allows coupling of the aid to the subject's ear via a conventional earmold. Although programmed using a computer, the WADHA is battery powered, using 4 AA batteries. Because the WADHA is fully digital, target gain values may be precisely specified at octave and inter-octave frequencies from 250 to 8000 Hz . This versatility in fitting allows for greater ease in matching hearing aid gain to many different hearing loss configurations, up to approximately 80 dB HL.

Figure 1. The CID Wearable Digital Hearing Aid (WADHA).



THE WADHA FEEDBACK EQUALIZATION ALGORITHM (FBE)

The Feedback Equalization (FBE) algorithm (Engebretson, et al, 1990), developed for the CID WADHA, suppresses feedback by adaptively cancelling the feedback path. The FBE algorithm is a continuously adaptive process that assumes the feedback path is constantly changing. Thus, the filter that is used to cancel the feedback is constantly being updated. Figure 2 shows a block diagram of the feedback path, and the FBE algorithm described in this study.

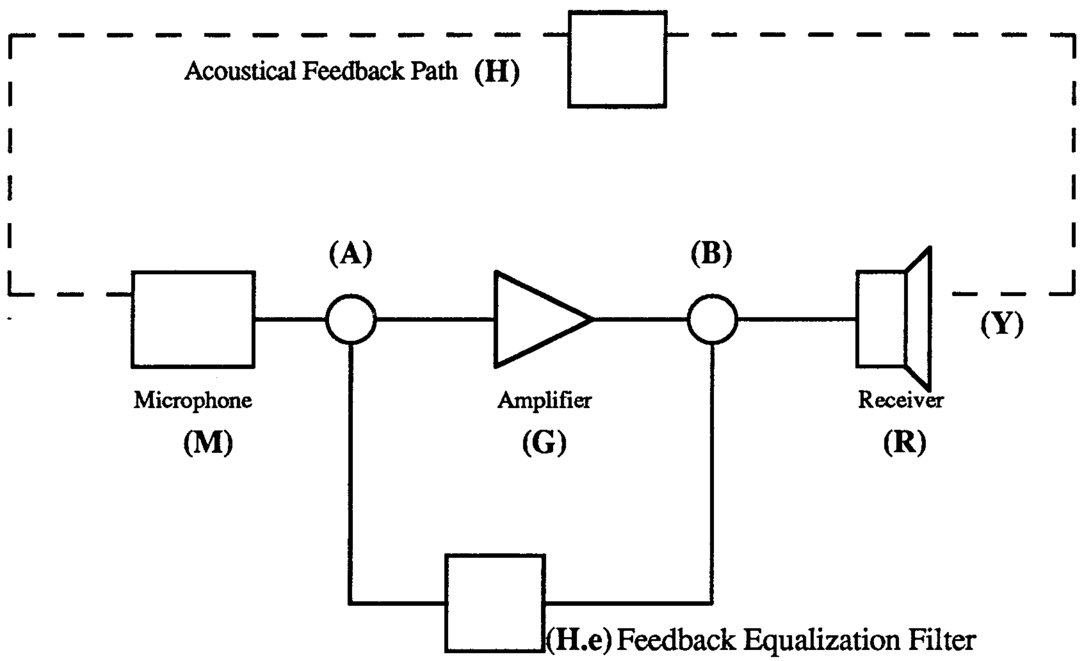


Figure 2. Model of the feedback pathway and the WADHA Feedback Equalization Algorithm.

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 Feedback equalization algorithm (H.e) monitors the processed signals from the hearing aid circuitry (G) at point (B), which includes both the input from the microphone and the hearing aid noise floor. This signal then travels to (R), the hearing aid receiver, which presents the acoustical signal to the ear canal. Some of this sound may escape and be carried back by the feedback pathway to the hearing aid microphone (M). The feedback equalization algorithm (H.e) monitors the input at (A) and looks for correlated signals between what was present at (B) and what is now present at (A). If the signals are correlated, then that correlated signal must have traveled from the receiver (R) through the feedback pathway, and represents acoustical feedback. The feedback equalization scheme compares (B) and (A), looks for a correlation, examines the difference between them, and adapts the filter (H.e) such that the feedback signals are cancelled.As the characteristics of the feedback path change (i.e., as with increased output from the hearing aid), the feedback equalization filter continuously adapts to the changing conditions to eliminate feedback pathway signals, while preserving desired (non-correlated) signals. 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 to 1.5 seconds to identify the feedback signals and adapt to them, however, any changes that occur in the feedback pathway are adaptively cancelled at a much faster rate (approximately 100 msec).

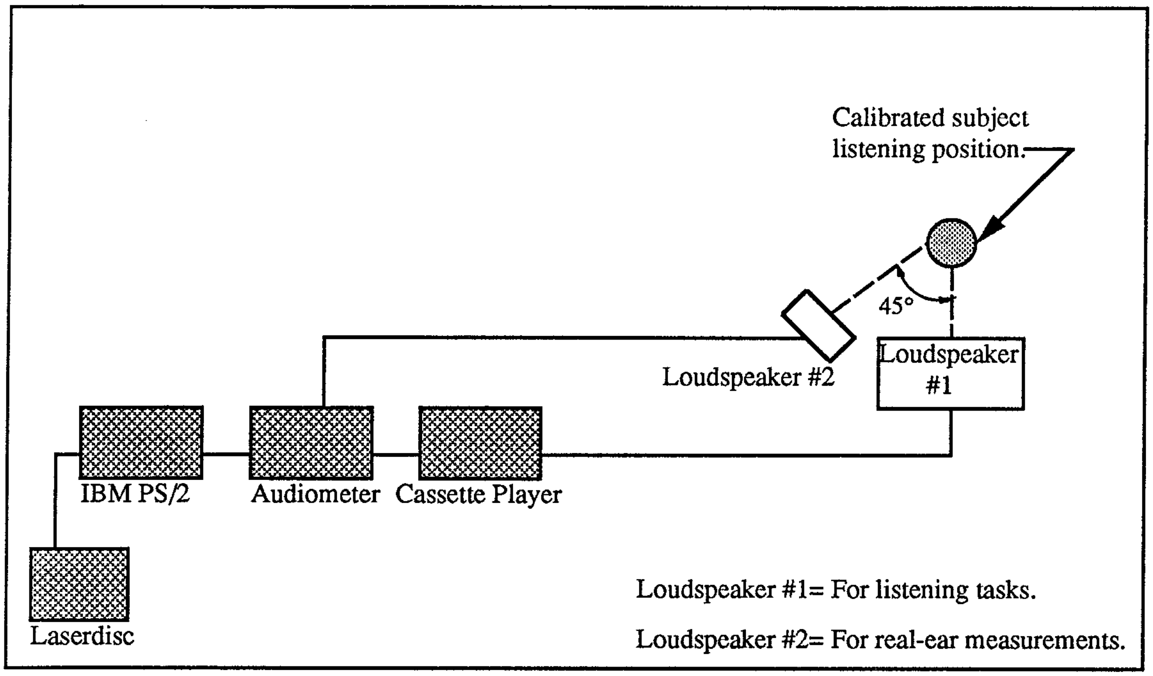


Figure 3. Diagram of room layout used for testing in this study.

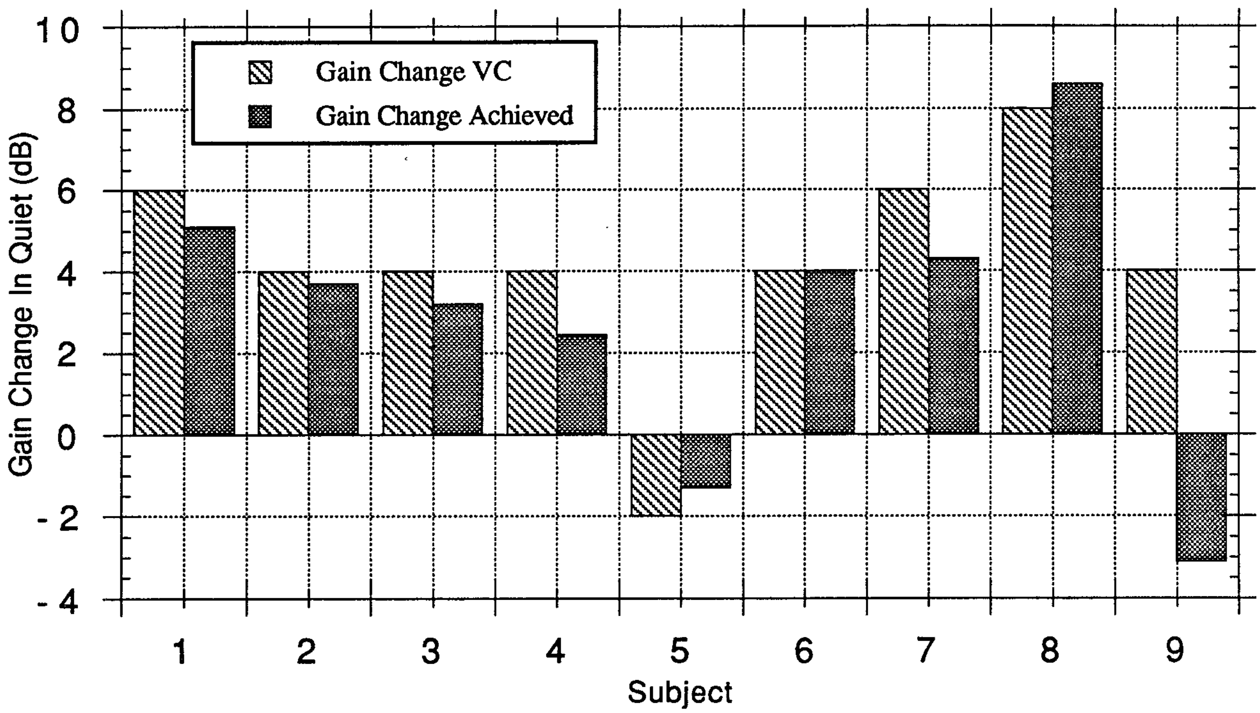


Figure 4. Distribution of gain changes made by subjects with FBE active while listening in quiet. Gain change VC refers to volume control changes, and Gain change achieved is real-ear gain change measured.

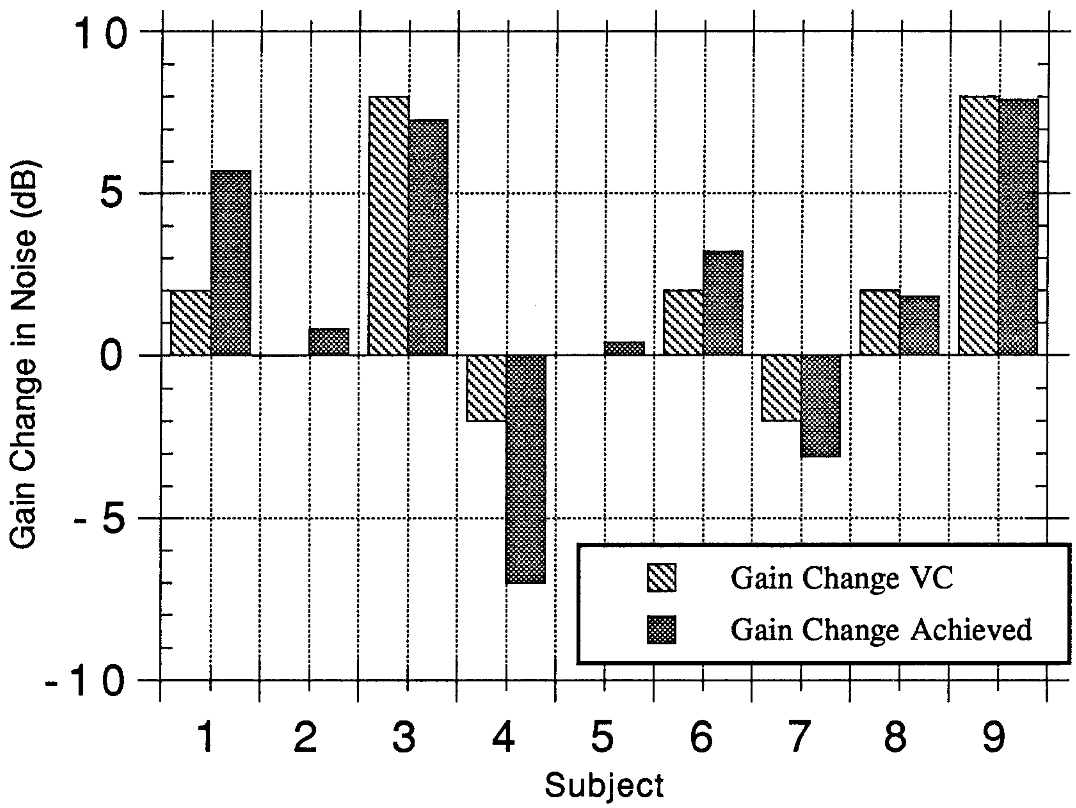


Figure 5. Distribution of gain changes made by subjects with FBE active while listening in noise. Gain change VC refers to volume control changes, and Gain change achieved is real-ear gain change measured.

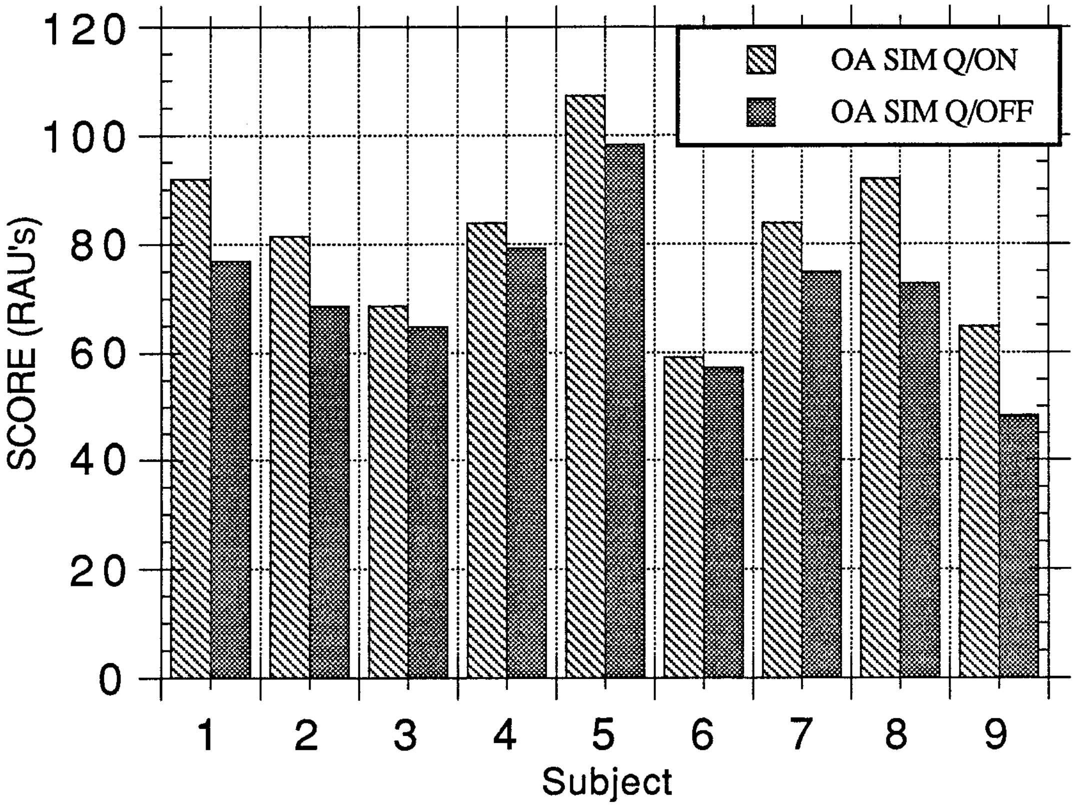


Figure 6. PHFWL scores for subjects with WADHA simulation of their own aid in quiet, with FBE ON and FBE OFF. Scores are converted from percentages to RAU's (see text).

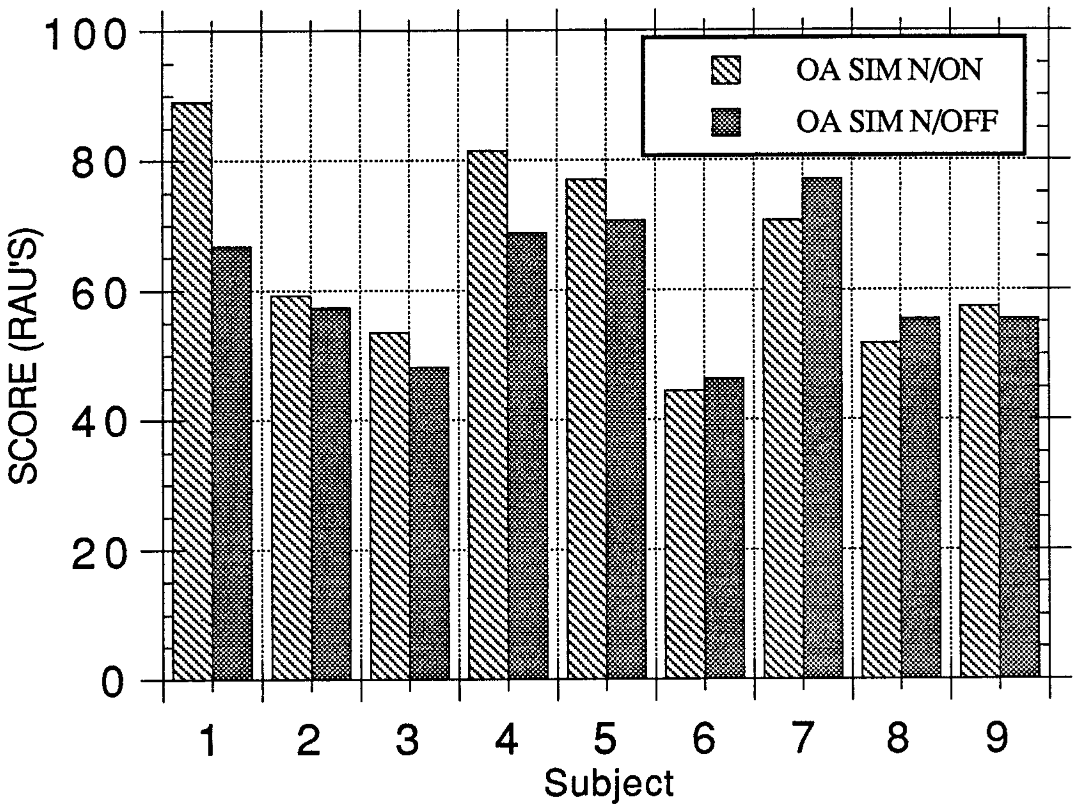


Figure 7. PHFWL scores for subjects with a WADHA simulation of their own aid in noise, for FBE OFF and FBE ON. Scores are converted from percentages to RAU's (see text).

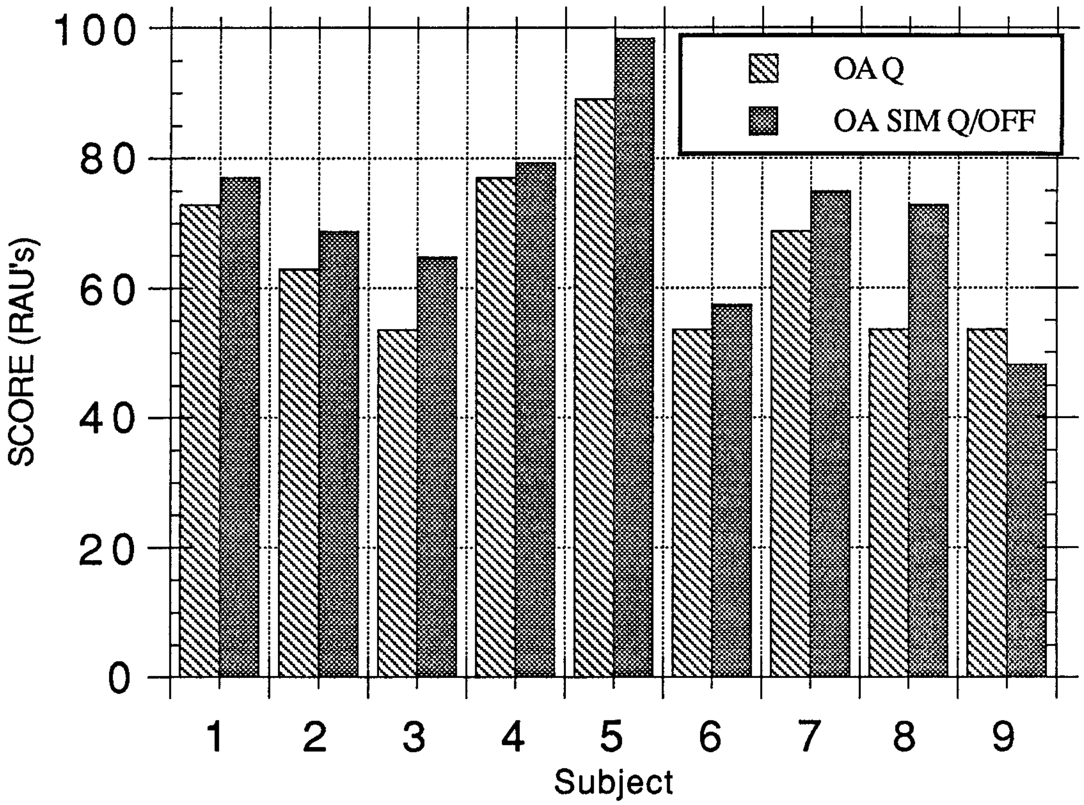


Figure 8. Comparison of subject's performance for listening with their own aid or a WADHA simulation of their own aid (FBE OFF) in quiet. Scores are converted from percentages to RAU's (see text).

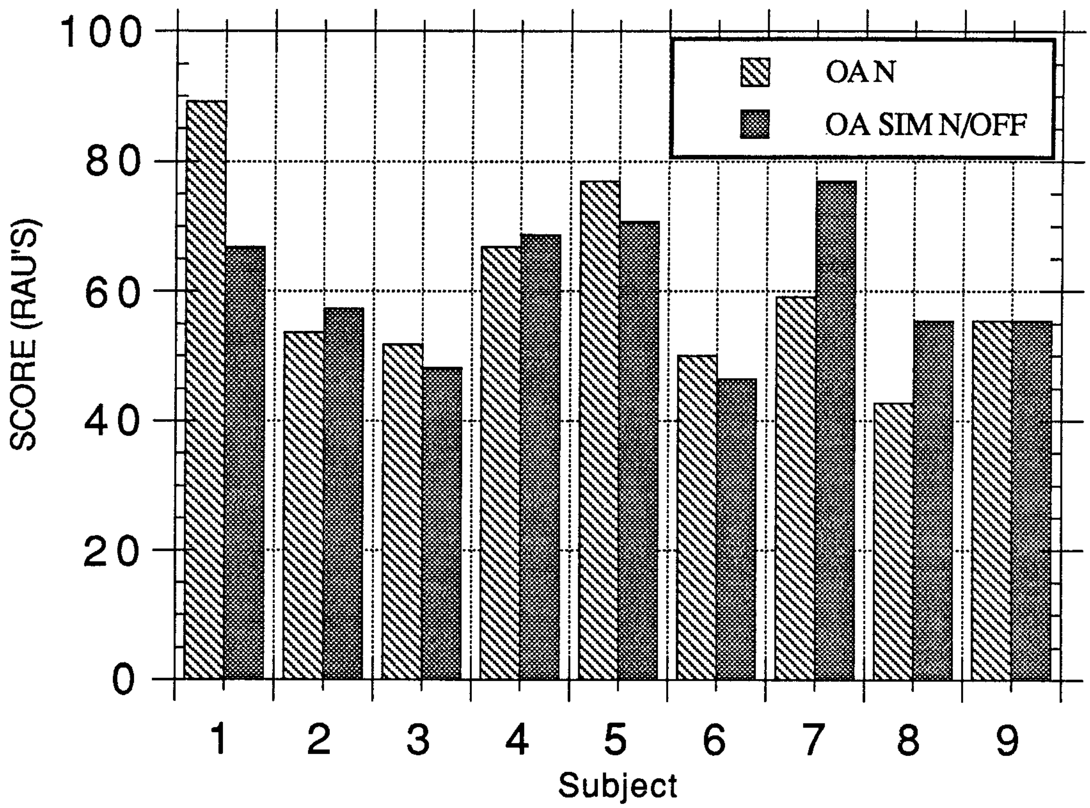


Figure 9. Comparison of subject's performance for listening with their own aid or a WADHA simulation of their own aid (FBE OFF) in noise. Scores converted from percentages to RAU's (see text).

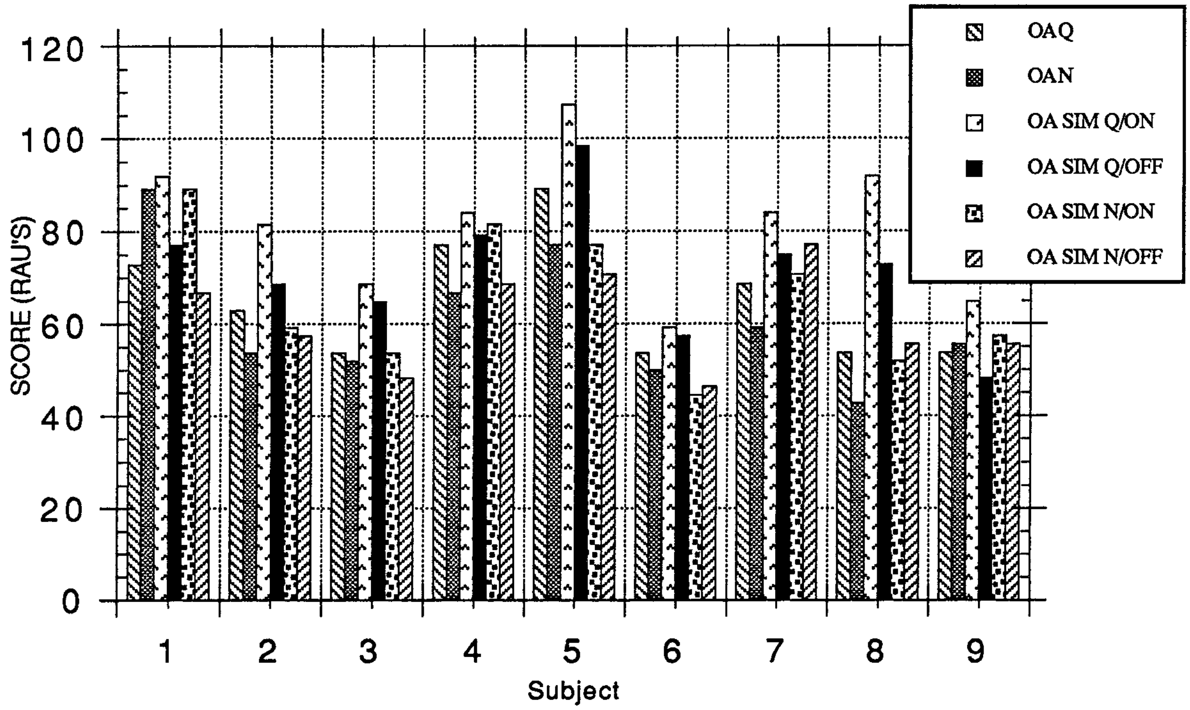


Figure 10. PHFWL scores for subjects in all listening conditions. Scores are converted from percentages to RAU's (see text).

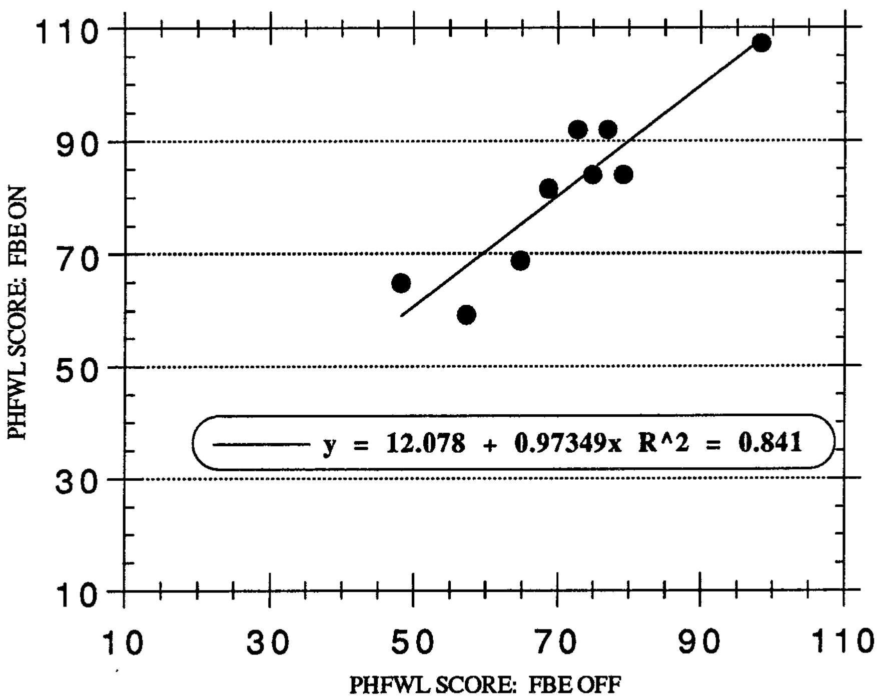


Figure 11a. Correlation between subjects' PHFWL scores, FBE ON vs FBE OFF, while listening in quiet to a WADHA simulation of their own aid.

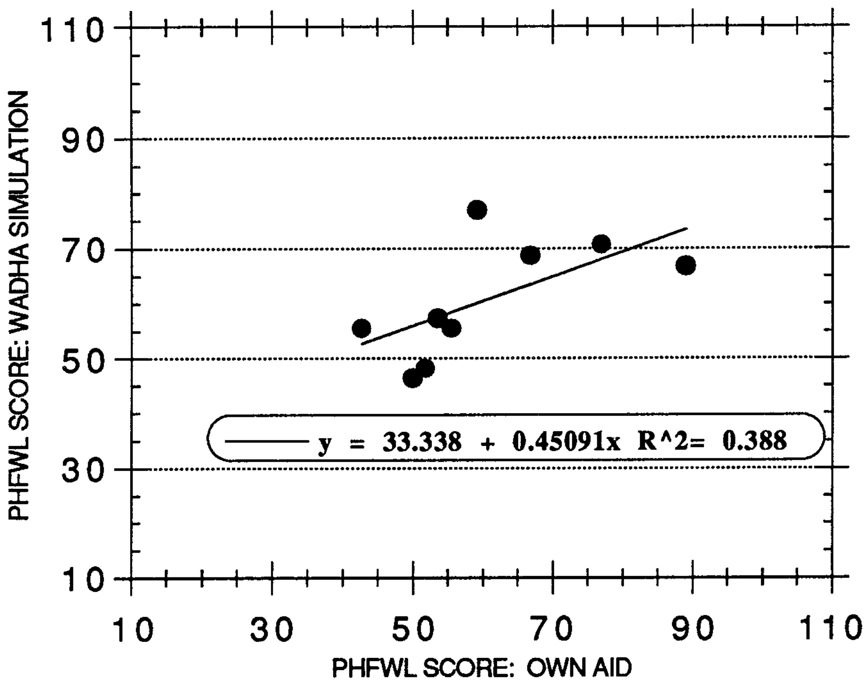


Figure 11b. Correlation between subject's PHFWL scores, WADHA simulation of their own aid and own aid while listening in noise.

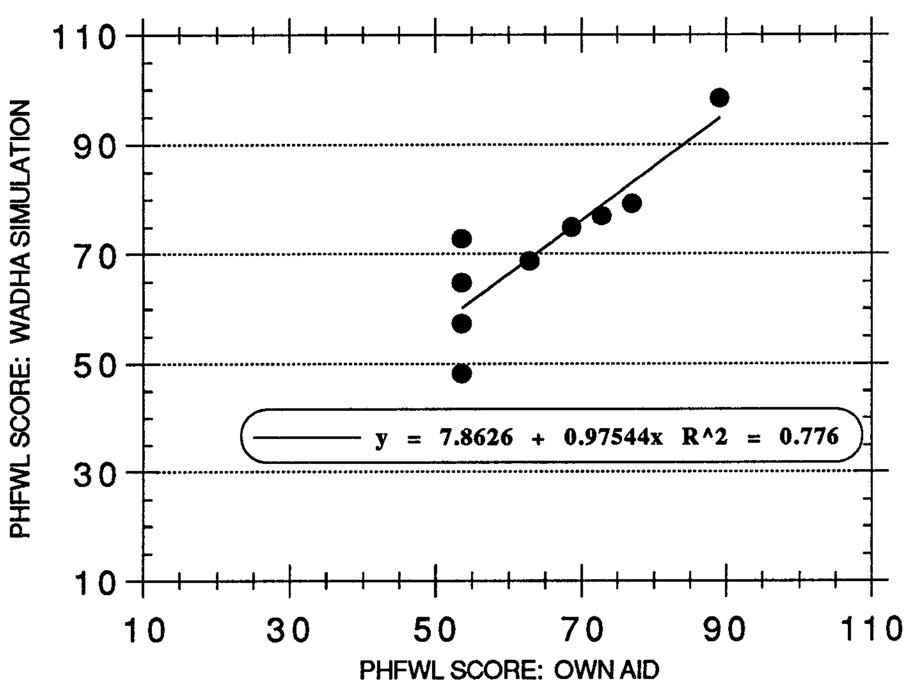


Figure 11c. Correlation between subjects' PHFWL scores, WADHA simulation of their own aid and own aid, while listening in quiet.

Table III. Accuracy of WADHA simulation of subjects' own hearing aid gain configurations, measured every 50 Hz across two bandwidths as Root-Mean-Square (RMS) error of fit.

	RMS Fit Error 250-6000 Hz				RMS Fit Error 500-2500 Hz			
Subject	Quiet ON	Quiet OFF	Noise ON	Noise OFF	Quiet ON	Quiet OFF	Noise ON	Noise OFF
1	6.0 dB	5.4 dB	7.0 dB	10.0 dB	7.4 dB	6.2 dB	9.4 dB	9.1 dB
2	5.0 dB	5.8 dB	5.3 dB	5.3 dB	4.5 dB	3.8 dB	4.7 dB	4.7 dB
3	3.1 dB	3.3 dB	4.2 dB	3.1 dB	1.7 dB	2.1 dB	3.1 dB	2.1 dB
4	7.8 dB	4.5 dB	9.2 dB	7.7 dB	4.8 dB	4.5 dB	7.0 dB	5.5 dB
5	4.9 dB	4.9 dB	4.8 dB	4.8 dB	4.2 dB	4.3 dB	3.6 dB	3.6 dB
6	2.5 dB	3.7 dB	2.5 dB	2.7 dB	1.3 dB	1.4 dB	1.1 dB	1.7 dB
7	3.8 dB	3.1 dB	7.4 dB	7.1 dB	1.8 dB	2.1 dB	5.1 dB	5.5 dB
8	2.4 dB	4.1 dB	2.3 dB	5.0 dB	1.7 dB	5.1 dB	1.3 dB	4.2 dB
9	3.2 dB	6.2 dB	5.0 dB	2.8 dB	2.8 dB	7.7 dB	3.7 dB	2.0 dB
	X=4.3 dB	**X=4.6 dB**	**X=5.3 dB**	**X=5.4 dB**	**X=3.4 dB**	**X=4.1 dB**	**X=4.3 dB**	**X=4.7 dB**
	X= 4.45 dB		**X=5.35 dB**		**X=3.75 dB**		**X=4.5 dB**	

ON = Feedback Equalization (FBE) active.

OFF= Feedback Equalization (FBE) inactive.

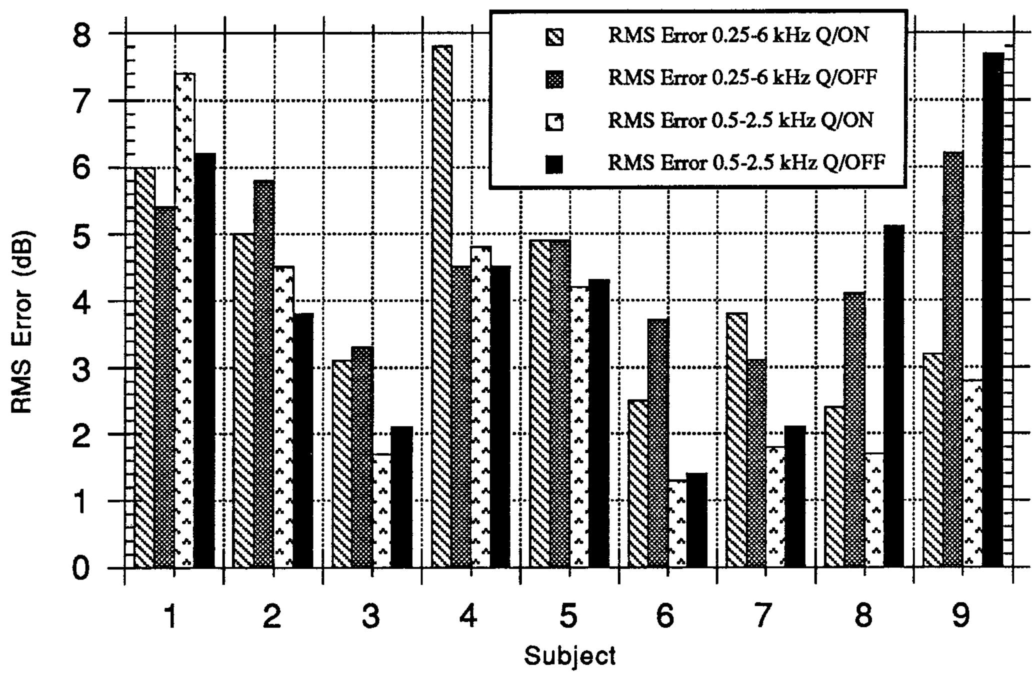


Figure 12. Accuracy of WADHA simulation of subjects' own hearing aid configurations, while listening in quiet, measured across two bandwidths as Root-Mean-Square (RMS) error of fit.

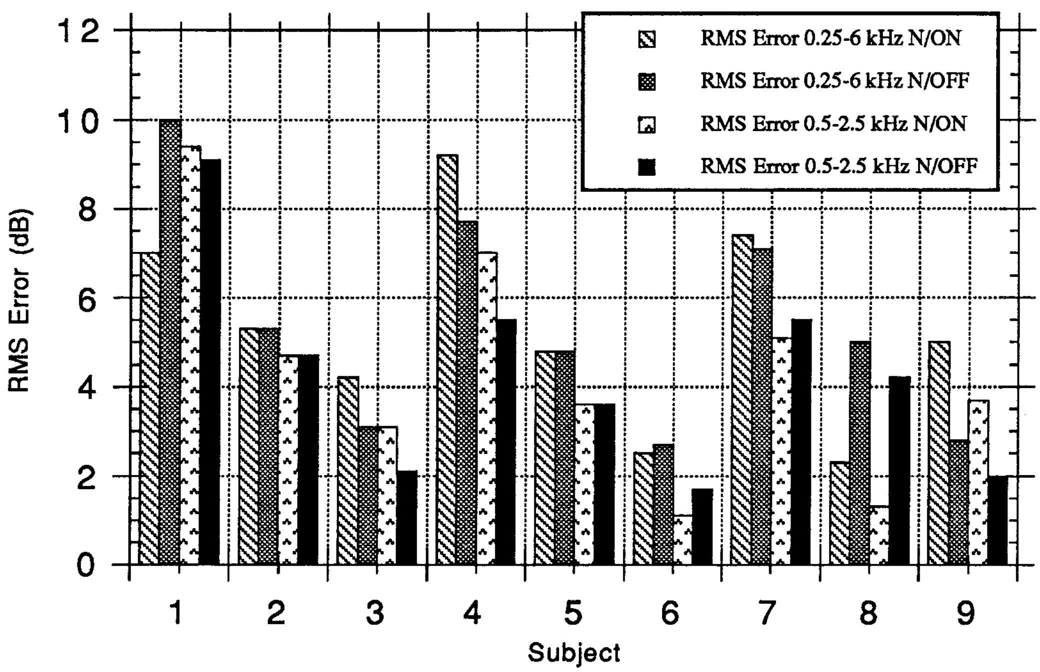


Figure 13. Accuracy of WADHA simulations of subjects' own hearing aid configurations, while listening in noise, measured across two bandwidths as Root-Mean-Square (RMS) error of fit.

Table IV. Gain functions derived from real-ear measurements of subjects' own aids, which were used as target gains for the WADHA simulations, in dB.

	Subject 1		Subject 2		Subject 3		Subject 4		Subject 5		Subject 6		Subject 7		Subject 8		Subject 9	
Frequency (Hz)	Q	N	Q	N	Q	N	Q	N	Q	N	Q	N	Q	N	Q	N	Q	N
250	26	32	1	5	1	1	7	21	4	12	29	35	19	7	5	3	18	18
500	45	47	6	6	2	3	19	21	9	14	37	39	26	19	10	10	21	22
750	50	53	12	13	3	8	35	34	18	24	39	42	32	25	13	15	32	32
1000	53	56	15	18	9	16	46	49	21	25	45	50	36	33	16	17	46	46
1500	45	46	20	25	16	25	42	41	17	20	49	50	37	35	24	26	39	39
2000	56	55	22	23	14	24	44	43	11	24	41	45	35	31	26	28	38	37
3000	26	33	12	13	9	13	16	21	18	22	40	42	37	33	12	14	20	19
4000	35	33	4	3	-5	10	13	20	11	18	24	33	28	25	19	18	21	23
6000	18	20	-6	-1	0	5	19	27	-11	1	12	17	22	14	8	1	11	11
X- 500, 1000, and 2000 Hz.	**51**	**53**	**14**	**16**	**8**	**14**	**36**	**38**	**14**	**21**	**41**	**45**	**32**	**28**	**17**	**18**	**35**	**35**

Q = Gain function while listening in quiet.

N = Gain function while listening in noise.

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