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# Individual Differences within and across Feedback Suppression Hearing Aids

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**Abstract**

**Background:** New and improved methods of feedback suppression are routinely introduced in hearing aids; however, comparisons of additional gain before feedback (AGBF) values across instruments are complicated by potential variability across subjects and measurement methods.

**Purpose:** To examine the variability in AGBF values across individual listeners and an acoustic manikin.

**Research Design:** A descriptive study of the reliability and variability of the AGBF measured within six commercially available feedback suppression (FS) algorithms using probe microphone techniques.

**Study Sample:** Sixteen participants and an acoustic manikin.

**Results:** The range of AGBF across the six FS algorithms was 0 to 15 dB, consistent with other recent studies. However, measures made in the participants ears and on the acoustic manikin within the same instrument suggest that across instrument comparisons of AGBF measured using acoustic manikin techniques may be misleading, especially when differences between hearing aids are small (i.e., less than 6 dB). Individual subject results also revealed considerable variability within the same FS algorithms. The range of AGBF values was as small as 7 dB and as large as 16 dB depending on the specific FS algorithm, suggesting that some models are much more robust than others.

**Conclusions:** These results suggest caution when selecting FS algorithms clinically since different models can demonstrate similar AGBF when averaging across ears, but result in quite different AGBF values in a single individual ear.

**Key Words:** Feedback suppression, hearing aid, individual differences

**Abbreviations:** AGBF = additional gain before feedback; FS = feedback suppression; KEMAR = Knowles Electronics Manikin for Acoustic Research; NAL-NL1 = National Acoustics Laboratory—Nonlinear 1; REAR = real ear aided response; SAV = select-a-vent

**Sumario**

**Antecedentes:** Métodos nuevos y mejorados de supresión de la retroalimentación se introducen rutinariamente en los auxiliares auditivos; sin embargo, la comparación de los valores de ganancia adicional antes de la retroalimentación (AGBF) en los diferentes instrumentos es complicada por la variabilidad potencial entre los sujetos y los métodos de medición.

**Propósito:** Examinar la variabilidad en los valores de AGBF entre sujetos individuales y el maniquí acústico.

**Diseño de la Investigación:** Un estudio descriptivo sobre la confiabilidad y variabilidad de la AGBF medidos en seis algoritmos comercialmente disponibles de supresión de la retroalimentación (FS) usando técnicas de micrófono sonda.

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**Muestra del Estudio:** Dieciséis participantes y un maniquí acústico.

**Resultados:** El rango de la AGBF en los seis algoritmos de FS fue de 0 a 15 dB, consistente con otros estudios recientes. Sin embargo, las mediciones realizadas en los oídos de los participantes y en el maniquí acústico con el mismo instrumento sugieren que las comparaciones de la AGBF entre instrumentos, usando las técnicas de maniquí acústico, pueden ser confusas, especialmente cuando las diferencias entre los audífonos son pequeñas (p.e., menos de 6 dB). Los resultados de sujetos individuales también revelaron una variabilidad considerable dentro de los mismos algoritmos de FS. El rango de valores de la AGBF fue tan pequeño como 7 dB y tanto como 16 dB dependiendo del algoritmo FS específico, sugiriendo que algunos modelos son mucho más robustos que otros.

**Conclusiones:** Estos resultados sugieren cautela cuando se seleccionan clínicamente los algoritmos de FS dado que diferente modelos puede demostrar una AGBF similar cuando se promedia entre los oídos, pero que resultan en valores de la AGBF muy diferentes en el oído de un sujeto individual.

**Palabras Clave:** Supresión de la retroalimentación, auxiliar auditivo, diferencias individuales

**Abreviaturas:** AGBF = ganancia adicional antes de la retroalimentación; KEMAR = Maniquí Electrónico Knowles para Investigación Acústica; NAL-NL1 = Laboratorios Nacionales de Acústica – No Lineal 1; REAR = respuesta amplificada de oído real; SAV = seleccione un orificio de ventilación

For the hearing aid wearer, the annoyance caused by feedback can drastically reduce patient satisfaction with the instrument (Kochkin, 2003). In fact, 28% of hearing aid wearers are reportedly dissatisfied with a device due to whistling, a common complaint associated with feedback (Kochkin, 2005). Acoustic feedback occurs when amplified sound from the receiver finds a pathway back to the microphone and is re-amplified by the hearing aid repetitively (i.e., a feedback loop). Acoustic feedback is almost always present; however, the level is often not high enough to result in oscillations that produce audible feedback. A hearing aid will generate audible feedback at any frequency that has a period equal to an integer multiple of the travel time required for sound to traverse the entire feedback and gain loop provided the hearing aid gain is greater than the feedback loop attenuation. The presence of audible feedback can therefore be reduced or eliminated either by increasing the distance between the receiver outlet and the microphone inlet or by acoustically isolating the receiver from the microphone (increasing feedback loop attenuation). Until recently, feedback was generally managed by the use of larger BTE (behind-the ear) instruments, which can lead to complaints related to cosmetics; or by limiting the vent size in smaller custom products or earmolds (Kuk, 1994), which can decrease comfort and increase occlusion. That is, the greater distance between the sound inlet and sound outlet in traditional BTE instruments decreases the susceptibility to feedback compared to smaller custom instruments, but the large case placed behind the ear combined with the need for tubing and an earmold sometimes leads to complaints about cosmetics. This feedback versus occlusion quandary has existed for many decades and often has been considered a major contributor to hearing aid dissatisfaction. Fortunately, the advent of digital feedback suppression (FS) algorithms for use in hearing aids provides a potentially effective technique to offset the feedback versus occlusion dilemma. Several digital signal processing based FS methods have been proposed and implemented in commercial hearing aids, and the refinement of such procedures continues at the time of this writing (e.g., Kates, 1999; Ji et al, 2005; Boukis et al, 2007; Lee et al, 2007). Effective FS algorithms can have considerable benefit clinically either by allowing for more gain while maintaining appropriate venting (potentially leading to improved audibility if the patient is underfit due to feedback limitations), or by allowing for increased venting without decreasing gain (potentially leading to improved comfort). The advent of effective FS algorithms led to the introduction of the modern open canal hearing aid style, which is enjoying increasing popularity (Johnson, 2006; Mueller and Ricketts, 2006).

Given that newer, and often times more effective, feedback suppression systems are regularly being introduced in commercial hearing aids, several authors have advocated methods for evaluating the effectiveness and limitations of FS in commercial hearing aids (e.g., Joson et al, 1993; Kaelin et al, 1998; Freed and Soli, 2006; Merks et al, 2006; Parsa, 2006; Shin et al, 2007). While the details of the methods proposed to date differ, the majority have described assessment of feedback suppression when the hearing instruments are fitted to an ear simulator mounted in an acoustic manikin such as the Knowles Electronics Manikin for Acoustic Research (KEMAR). Feedback is usually then initiated by some combination of increasing gain, increasing venting, and/or introducing a physical surface near the hearing aid. The potential benefits and limitations of activating FS algorithms are then quantified in several ways, including (1) the magnitude of additional gain available before feedback,

commonly referred to as *gain margin*, *added stable gain*, *added gain before feedback*, or a host of other terms; (2) the duration required to eliminate feedback after introducing a physical object, referred to herein as *feedback adaptation time*; (3) the misidentification of an external signal as feedback and introduction of artifact or distortion by attempting to cancel this signal, commonly referred to as *entrainment*, *maladaption*, or a number of other terms; and (4) the introduction of other signal artifacts or distortions.

Although it is clearly important that the FS algorithm does not introduce unwanted side effects, the most commonly discussed attribute related to performance is the magnitude of additional gain available before feedback. For the purposes of this manuscript we operationally define the magnitude of additional gain available before feedback (AGBF) as the REAR (real ear aided response) values measured 2 dB below audible feedback with the feedback suppression system activated minus those measured 2 dB below audible feedback with the feedback suppression system deactivated, *while maintaining the same frequency response shape*. This term and definition can be contrasted with the commonly used term *added stable gain* in that the stability of the system is not directly measured, instead AGBF only considers a fixed point below audible feedback. Despite these differences, AGBF is expected to be nearly identical in magnitude to added stable gain.

Since new and improved methods of feedback suppression are routinely introduced, it is not surprising that reports using acoustic manikin-based measures of feedback suppression have shown clear and large differences across commercial hearing aids. Added stable gain values ranging from 0 dB to more than 18 dB have been reported (e.g., Freed and Soli, 2006; Merks et al, 2006; Shin et al, 2007). What is unclear from these studies, however, is whether individual patients obtain similar additional gain before feedback from the same FS system. From a clinical perspective it is useful to measure the variability across individual listeners with regard to FS. For example, if an AGBF value of 10 dB is reported for a specific instrument, can I expect all my patients to receive this same magnitude? If not, how much variability is expected, and is this variability affected by factors such as the hearing loss configuration or the specific FS algorithm selected? Answering these questions is important in order to set realistic expectations related to hearing aid fitting range and AGBF values in individual patients.

AGBF is expected to differ as measured in individual ears based on a number of factors. First, some variation might be expected due to measurement error. Specifically, a number of authors have shown that repeated real ear gain measures using probe microphone equipment can reveal standard deviations in the 1500–3000 Hz frequency range of between approximately 0.5 and 3 dB, depending on the measurement method (e.g., Ringdahl and Leijon, 1984; Dillon and Murray, 1987; Killion and Revit, 1987; Barlow et al, 1988). Standard deviations of less than 1 dB are generally reported when the loudspeaker is placed at a 45° azimuth (e.g., Killion and Revit, 1987) as compared to 0° azimuth. However, of greater interest for the current study, a number of other factors such as individual differences in ear canal geometry and the specific hearing aid to ear coupling may affect the feedback loop resulting in differences in susceptibility to feedback (e.g., Kates, 1988; Egolf et al, 1989; Hellgren et al, 1999; Rafaely et al, 2000). Since accurate modeling of the feedback loop by the FS algorithm is crucial for optimal feedback suppression, the magnitude of AGBF is expected to differ across patients. Further, it is speculated that there will be variability in the robustness of commercially available FS algorithms with regard to individual differences. If so, the range of AGBF values measured within a group of individuals would be expected to differ across manufacturers.

Only a few studies have formally examined AGBF (or added stable gain) across FS algorithms using actual hearing impaired subjects and realistic hearing aid coupling systems (Greenberg et al, 2000; Johnson et al, 2007). Neither of these investigations compared the AGBF as measured in actual patients to those measured using an acoustic manikin. In addition, both studies fitted patients according to their individual hearing loss configurations. While it is clearly important to fit to a range of real hearing loss configurations to obtain average results that are indicative of actual clinical fittings, such a strategy somewhat confounds interpretation of individual differences. Specifically, it is unclear whether differences measured across individuals were due to differences in how the algorithm interacted with the specific ear or how it interacted with the gain configuration (or both).

The purpose of this study was twofold. First, it was of interest to examine the AGBF afforded to individual listeners within and across five commercially available hearing aids. All participants were fitted with the same two hearing aid gain configurations in order to differentiate between the effects of gain configuration and individual differences related to ear geometry and/or coupling on the operation of the FS algorithm. Secondly, it was of interest to compare the values measured in real ears to those obtained using a KEMAR with regard to magnitude and variability.

## METHODS

### Pilot Measures: Venting

Prior to the main study, the test procedures described below were evaluated using three individuals with

normal hearing. For this testing, each hearing aid was coupled to the ear using a custom-fit skeleton earmold with size 13 tubing and a 3 mm select-a-vent (SAV), as well as an open canal fit. The open canal fit was completed using standard #13 tubing placed in the ear canal using the standard "fit-n-go" open canal fitting kit that is available from Phonak AG. The 3 mm SAV was selected as the largest size that was possible when considering all three participants individual ear sizes according to the earmold manufacturer. AGBF was evaluated for a 2 mm vent plug, the unplugged 3 mm vent, and tube fit conditions. This testing revealed that, for the two venting conditions, gain could be increased to the amplifier maximum for both hearing loss configurations across all three participants for three of the five hearing aid brands (listed as Brands A, B, and E in this manuscript). That is, for these brands and test conditions, amplifier gain rather than feedback limited the measured AGBF. Since a primary goal of this research was to examine individual differences in AGBF across hearing aid brands due to the potential interaction between the FS algorithm and ear geometry, the decision was made to complete all further testing using only the open canal fit. With this procedure, each participant's measured AGBF values were limited by the effectiveness of the FS algorithm rather than the hearing aid's available amplifier gain.

### Participants and Hearing Aids

A total of 16 adult hearing aid wearers were fitted with a total of five commercial hearing aids. One of these models (Brand C) had two separate FS algorithms available, and it was tested using both (referred to as *Cf* and *Cs* for fast and slow adaptation times, respectively). All participants had at least a moderate sloping sensorineural hearing loss (as identified by clinical records), but the degree of hearing loss is not reported here since no behavioral or subjective data were gathered as part of this investigation. Participants with hearing loss were recruited as part of a series of studies that also included collection of behavioral data. All measurements were completed in each participant's right ear. It was speculated that hearing loss configuration might affect the magnitude of AGBF across FS algorithms. However, it was of interest to examine the potential affect of hearing loss configuration while limiting the number of variables in order to also systematically assess the variability of AGBF as a function of ear geometry. Therefore we chose to preselect two hearing loss configurations and fit all subjects with the same two frequency response shapes as described below.

The five hearing aids brands were all standard BTE models (not the miniature BTE models used for many companies' current open canal products) from five leading manufacturers. Each model represented the manufacturer's "premiere" or most advanced hearing aid model as of May 2006. According to manufacturers' advertisements, all five models incorporated FS processing that modeled the feedback loop response with an adaptive filter, estimated the feedback signal, and then subtracted this estimated signal from the hearing aid input (commonly referred to as *feedback cancellation*).

Depending on the specific hearing aid, digital noise reduction processing can interact with the magnitude of hearing aid gain, potentially affecting the magnitude of additional gain before feedback. According to manufacturer literature, some feedback suppression algorithms measure the feedback loop response separately in omnidirectional and directional modes, while others only measure this response in omnidirectional mode. Therefore, all special features other than feedback suppression, including noise reduction and directional microphones, were disabled for testing. It also has previously been shown that the compression ratio may influence the adaptation speed of some phase FS algorithms and that measurements with fluctuating systems may depend on compression time constants that cannot be exactly matched across all hearing aid models (Greenberg et al, 2000). Unfortunately, not all hearing aid models allow for frequency shaping when set to linear mode, preventing an appropriate comparison of output margin when set to linear amplification. Given these limitations and our interest in focusing on the additional gain before feedback as measured in typical clinical fittings, we matched the selected models as closely as possible in terms of gain and compression parameters, and we simply acknowledge the limitation of differences in compression time constants and compression thresholds across the hearing aids tested in this study.

The two frequency response targets were calculated for a flat 50 dB HL hearing loss (250 through 8000 Hz) and a sloping hearing loss with thresholds assumed to be 20 dB HL at 250 Hz, 30 dB HL at 500 Hz, 50 dB HL at 1000 Hz, and 65 dB HL at 2000 through 8000 Hz. These two targets were calculated using the National Acoustic Laboratory—Non-linear 1 prescriptive method (NAL-NL1; Dillon, 1999). Since no behavioral data were gathered, and in order to further limit variability, the five hearing aid models were matched to target gain values in a 2 cc coupler. NAL-NL1 coupler targets were matched as closely as possible for 50 and 80 dB SPL input levels, while also attempting to match compression thresholds and ratios. The average 2 cc coupler output in response to a 50 dB SPL input across the five hearing aid brands is shown in Figure 1. These data generally reveal that we were successful in matching hearing aid gain for the specific test conditions. It was not possible to match instruments in terms of compression time constants since there was generally no control

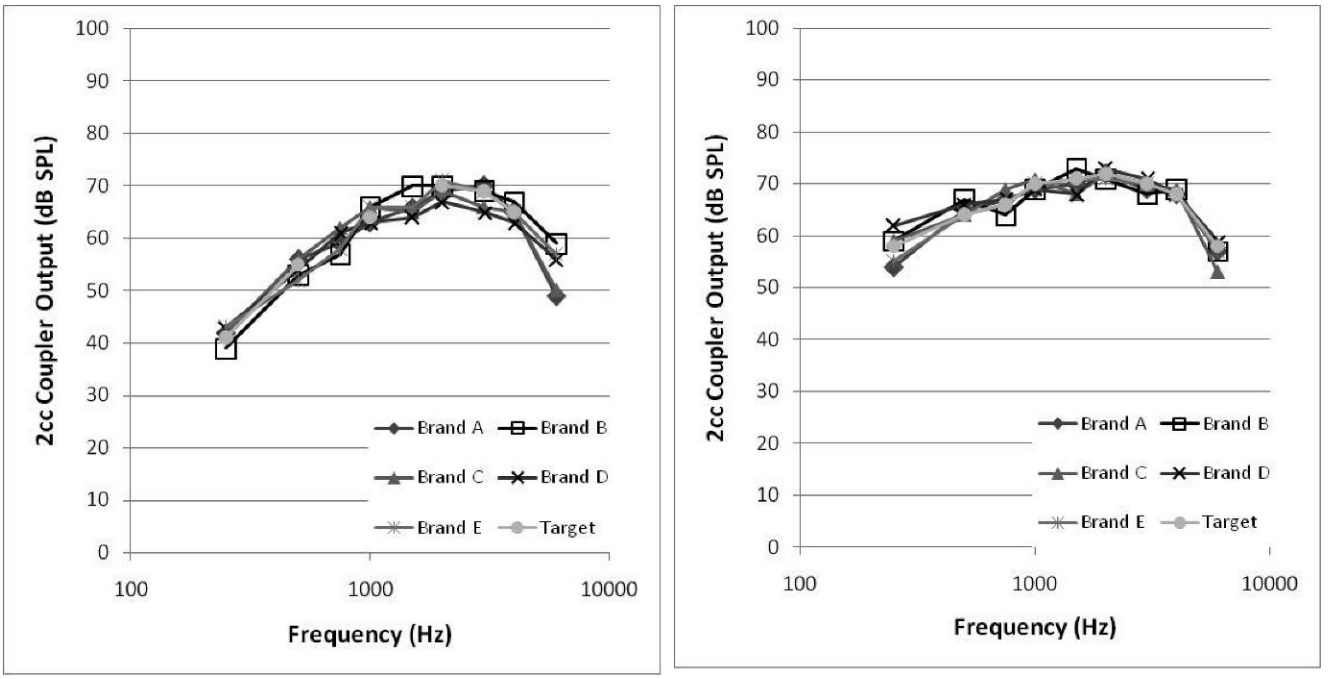


**Figure 1.** The 2-cc coupler match to NAL-NL1 targets for the five hearing aid models. The match to the "sloping" targets is shown in the left-hand panel, and the match to the "flat" targets is shown in the right hand panel.

provided for these parameters. The "shaped" real speech signal used with the Audioscan Verifit hearing aid test system ("carrot" passage) was used to match all target gain values. Special attention was paid to matching 50 dB SPL input targets since all subsequent probe microphone evaluation of the FS algorithms were completed in the presence of this same passage presented at 50 dB SPL in the sound field.

## Evaluation Procedures

Each participant was seated in a (3.6M × 4M × 2.7M) double walled sound booth in front of an Audioscan Verifit probe microphone system. Participants were seated 1.25 m from the Verifit loudspeaker in a cloth office chair with their arms to the side, placed on the chair armrests. Each of the five hearing aid models was evaluated randomly in turn, with Brand C evaluated in two different settings as described previously. The FS was first deactivated, and the hearing aids were placed on the patient using the gain settings that were previously programmed in the 2 cc coupler. It should be noted that because of this procedure, individual patients were not expected to receive an exact match to NAL-NL1 real ear targets for the specified hearing loss. Instead, the amount of real ear gain was expected to be matched across models within each patient. The REAR was then measured in response to the "shaped" male real speech signal (Verifit "carrot passage") presented at 50 dB SPL. The 50 dB SPL speech signal was played continuously during all testing with the goal of maintaining the same gain across all hearing aids and test conditions. A stored equalization method was used for all REAR measurements since real-ear measurements using the modified pressure method with concurrent (real-time) equalization can be inaccurate when using an open canal fitting (Mueller and Ricketts, 2006; Lantz et al, 2007). Specifically, amplified sound leaks out of the ear canal and reaches the reference microphone, resulting in a reduction of the output of the system loudspeaker to maintain the desired level. The magnitude of this error generally increases with the effectiveness of the FS system (Lantz et al, 2007), potentially increasing the apparent magnitude of the AGBF.

In order to more closely emulate clinical procedures, an attempt was made to maintain the same frequency response throughout testing. This was accomplished by adjusting the overall gain of each hearing aid model in 1 dB steps rather than adjusting gain in a channel specific manner. This procedure was intended to circumvent artificial inflation of the AGBF by not averaging frequencies at which considerable additional gain is present; however, feedback is unlikely (e.g., 500 Hz) with those frequencies for which maximum gain is more likely to be limited by feedback (e.g., 2000 Hz).

Audible feedback was identified by the experimenter, who monitored the probe microphone output using headphones. The experimenter had normal hearing (thresholds no poorer than 10 dB HL at all audiometric frequencies). While the use of this subjective procedure

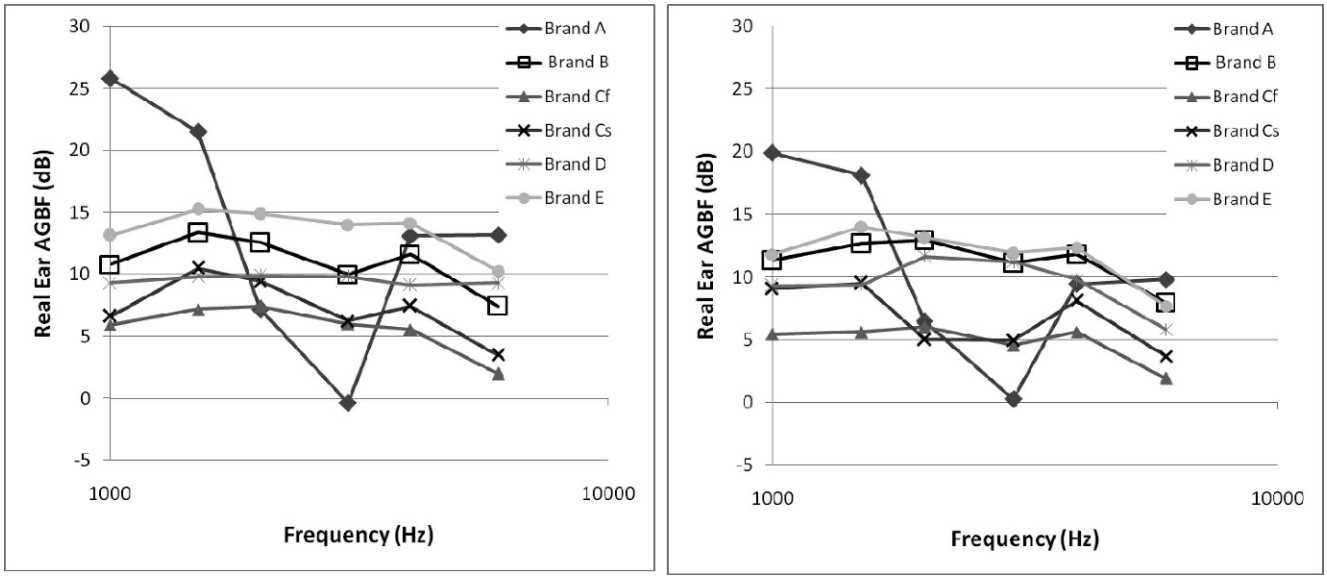


**Figure 2.** The average AGBF at ½-octave center frequencies measured in the ear of 16 hearing aid wearers across five commercially available hearing aids as measured using the Audioscan Verifit. All participants received the same two hearing aid gain configurations, which were programmed in a 2-cc coupler using targets for the same sloping (left-hand panel) and flat (right-hand panel) hearing loss configurations.

might be questioned because of the potential introduction of variability, it was selected for two reasons. Most importantly, since complaints related to feedback typically are related to hearing feedback, it was felt that this method offered the best face validity. Secondly, it is important to note that since the level of feedback does not exhibit linear growth with increases in gain, this method is much less subjective than a typical auditory threshold measure. That is, the feedback level rose from inaudible levels to clearly audible levels (more than 60 dB SPL as measured in the ear canal of the experimenter) over gain increases of 1 to 2 dB in the tested instruments. In pilot testing the experimenter was asked to identify the presence of feedback while blinded to the real ear output and hearing aid controls. This testing revealed gain settings within the same instrument never varied by more than 1 dB, suggesting excellent reliability.

With the FS system deactivated, hearing aid gain was increased and decreased in 1 dB steps to identify the overall gain level deemed 2 dB below the level that caused persistent feedback, and an REAR measure was taken. Each participant was asked to open and close his or her mouth to determine if this resulted in feedback at the 2 dB down from static feedback level; however, no additional feedback was identified even with jaw movement under the tested conditions. The FS algorithm was then activated. For the tested brands, the manufacturer-specified feedback algorithm initialization procedure was followed prior to activation. This initialization procedure varies across manufacturers and is proprietary in nature, but it is expected to be used to model the feedback loop response and/or limit the maximum available gain. After FS activation, overall gain was again increased, and the overall gain level that was 2 dB below feedback was again identified using the procedure just described with one minor deviation. This deviation consisted of providing up to five seconds of time for the hearing aid to eliminate feedback once audible. The difference between the two REAR measures (FS active minus FS inactive) was taken as the measure of AGBF. These procedures were conducted using both the flat and sloping hearing loss gain configurations for each of the five hearing aid models (with Brand C evaluated under two FS settings).

The exact procedures used with the experimental participants were also completed using an IEC 711artificial ear simulator (GRAS Type RA0045) mounted within a KEMAR. Specifically, the probe microphone was inserted in the ear simulator, the hearing aid was placed over the ear, the sound tube was placed in the ear simulator, and the measures were made as described above. In order to obtain a reliable measure, four repetitions of these measures were completed for each instrument. Each hearing aid was removed and reinserted between each measure.

## RESULTS

The AGBF for the five hearing aid brands for the sloping and flat hearing loss configurations are shown in Figure 2. Due to the limited low frequency gain available in open fitting configurations and the amplifier receiver limits in the highest frequencies, only 1000 through 6000 Hz are shown in these figures. The data in Figure 2 support that our goal to maintain

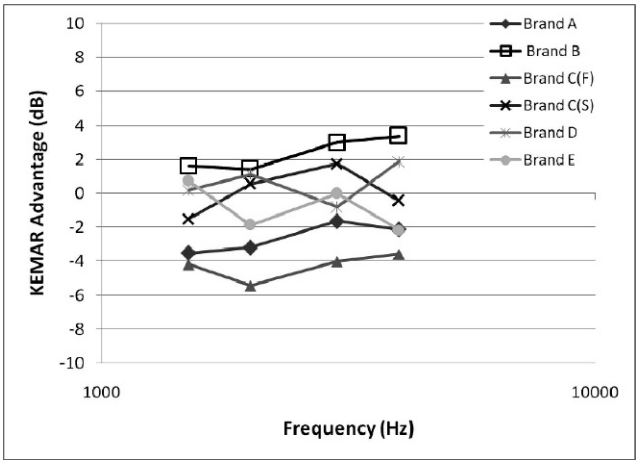


**Figure 3.** The difference between AGBF measured on the KEMAR and in 16 participants' ears. Positive values indicate that a larger AGBF value was measured in the KEMAR.

the same frequency response shape while increasing the gain was generally met for four of the five brands. That is, the gain margin was generally constant as a function of frequency within hearing aid Brands B, C, D, and E. The most common deviation from the flat AGBF function across these models is seen at 6000 Hz, where the amplifier/receiver limited the maximum gain available. A minor deviation is also visible in the middle frequencies, particularly for Brands B and Cs. While the underlying cause of this deviation is unknown, it is speculated that these brands may incorporate some kind of notch filter or gain limiter in addition to the feedback canceller as a secondary method of feedback suppression. Figure 2 also clearly reveals that the frequency response shape for Brand A did not remain constant despite the fact that only overall gain was adjusted. Unlike the other four models, feedback never occurred in the Brand A instrument with the FS algorithm activated, and the values in Figure 2 represent AGBF that was limited by the maximum hearing aid amplifier gain. The Brand A feedback initialization program was clearly used to limit the maximum hearing aid gain in some frequencies to completely avoid feedback. Unfortunately, this limiting also resulted in averaged AGBF values of approximately 7 dB at 2000 Hz and 0 dB at 3000 Hz. In other words, gain could not be increased above the original target values if it was also of interest to maintain the same frequency response shape.

For the purposes of further data analysis, the AGBF was collapsed into a single value representing the average from 1000 through 3000 Hz for Brands B, C, D, and E. Given our operational definition of AGBF, no averaging was completed for Brand A, since this value was always limited by the locking of gain at 3000 Hz. Therefore, only the AGBF measured at 3000 Hz, rather than an across frequency average, was used when comparing the AGBF of the Brand A instrument to the averaged AGBF of the other four hearing aid models.

Differences across averaged AGBF values were evaluated using a two-factor ANOVA. For this analysis the within-subjects factors were the six hearing aid FS algorithms (Brands A, B, Cs, Cf, D, and E) and the two hearing loss configurations (flat, sloping). Statistical significance was defined at $\alpha = 0.05$ level, and Tukey honest and significant difference (HSD) testing was used for post hoc analyses of the data. Results revealed a significant main effect of FS algorithm ($F_{2,5} = 790.4$, $P < 0.0001$). Post hoc analysis of this main effect revealed the AGBF was significantly less for Brand A than all other FS algorithms ($P < 0.0001$). Further, the average AGBF was significantly greater ($P < 0.002$) for Brand E than all other FS algorithms except Brand B (Brand E not significantly different from Band B). Finally, the average AGBF was significantly greater ($P < 0.003$) for Brand B than all other FS algorithms except Brands D and E (Brand B not significantly different from Brand D or E). No other significant effects or interactions were present. This lack of significant differences supports similar average AGBF across the two hearing loss configurations. The conclusion that AGBF was not significantly affected by gain configuration was further supported by examination of individual differences. Specifically, across all 96 comparisons (16 participants $\times$ 6 FS algorithms) there were only nine instances for which the difference in AGBF measured between the two gain configurations exceeded 3 dB.

The difference between the average of four repetitions of the AGBF measured in the KEMAR and the average measured for 16 individual participants is shown in Figure 3. Since significant differences were not found as a function of hearing loss configuration, these data represent the average AGBF values across both hearing loss configurations. These results show that the difference in AGBF between the KEMAR and real ears was less than 6 dB. However, for some models, greater AGBF was found on the KEMAR (e.g., Brand B), while for other models greater AGBF was found in real ears (e.g., Brands A and Cf). Finally, Brands Cs, D, and E revealed good agreement between KEMAR and real ear measures, with deviations of less than 2 dB.

## DISCUSSION

While not a primary goal of this experiment, the data shown in Figure 2 reveal a large range in the average AGBF provided across the six FS algorithms evaluated. This range (approximately 0 to 15 dB) is similar to the range of added stable gain values, which were previously reported by Freed and

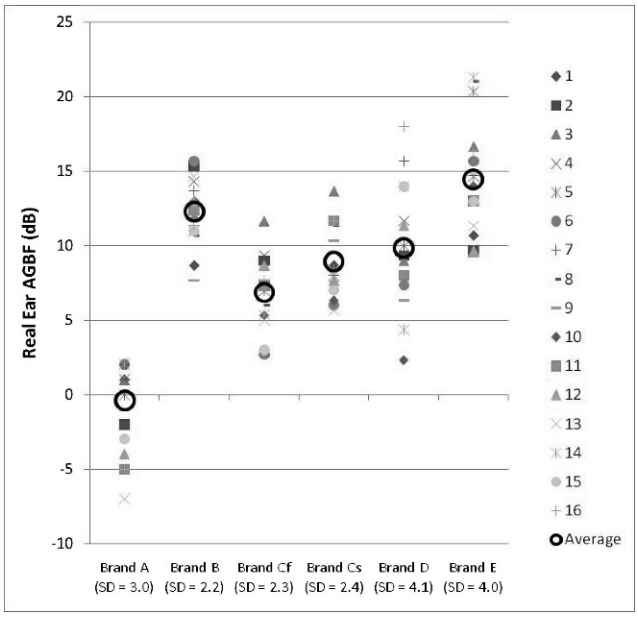


**Figure 4.** The AGBF measured in 16 participants fitted with a "sloping" gain configuration, averaged from 1000 to 3000 Hz (1/2-octave center frequencies), for the 6 FS algorithms evaluated.

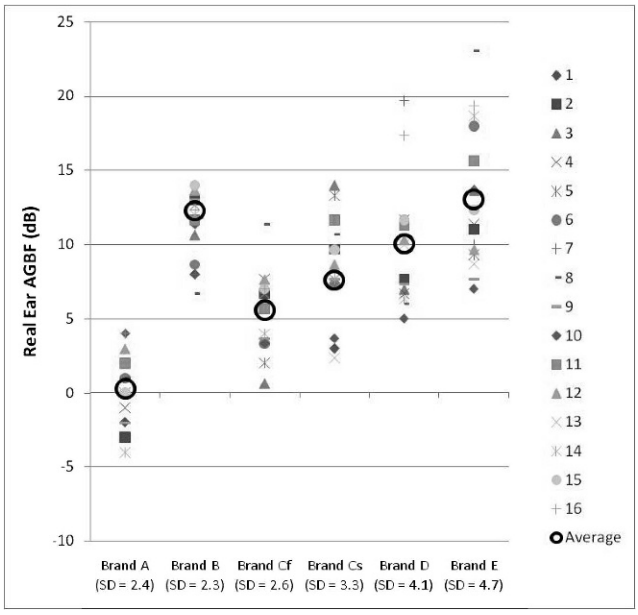


**Figure 5.** The AGBF measured in 16 participants fitted with a "flat" gain configuration, averaged from 1000 to 3000 Hz (1/2-octave center frequencies), for the 6 FS algorithms evaluated. The SD values represent one standard deviation for each FS algorithm.

Soli (2006), who examined nine commercially available hearing aid models (approximately 0 to 18 dB). This similarity occurred despite the fact that only an open canal coupling was used in the current study and the frequency response was held constant across gain settings and ears. Given the similarities between these two data sets, it was not surprising that there was reasonably good agreement between the KEMAR and real ear measures of AGBF in the current study (Figure 3). However, although differences between these two measures were generally small, it is proposed that using KEMAR-based measures to compare AGBF in commercially available FS algorithms may lead to specious results. The largest across brand difference was noted between Brands B and E (Figure 3). Based on the KEMAR data, it might be erroneously concluded that Brand B actually has AGBF values that are approximately 3 dB larger than Brand E, even though the real ear data reveal the opposite relationship (Figure 2). Consequently, we propose that KEMAR-based measures should only be used to obtain a rough estimate of the AGBF expected in real ears.

One main goal of this experiment was to examine the variability in AGBF in real ears. As pointed out by Rafaely et al (2000), the feedback loop response in human ears can be affected by three different factors, including (1) a change in how much sound leaks out of the ear; (2) a change in the sound-pressure level inside the ear canal (due to differences in ear canal impedance and/or geometry); and (3) a change in the level of the leaked feedback signal to the microphone due to the presence of reflective surfaces.

In the current study, an open canal fitting was evaluated without the presence of reflective surfaces. That is, the first and third factors listed above were expected to be relatively constant across participants and test conditions. This design allowed for a focus on whether ear geometry and ear canal impedance affected AGBF across the tested FS algorithms. In this experiment, individual participant differences within and across these six FS algorithms was therefore of particular interest. The results shown in Figure 2 revealed that the hearing aid gain configuration did not significantly affect the measured AGBF. However, it should be noted that only two gain configurations were evaluated (flat and sloping), and it is not known if some interaction between AGBF and gain configuration may have existed for more divergent configurations. Given the lack of a significant effect of hearing loss, and in order to examine potential individual differences as a function of FS algorithm, the individual subject data that were used to obtain the averaged data in Figure 2 were replotted in Figures 4 and 5.

While modern FS algorithms have been developed with robust stability across ears and listening environments as a design goal, Figures 4 and 5 demonstrate a large range of AGBF in individual listeners fitted with the same FS algorithm. Further, the magnitude of variability across participants differed depending on the specific FS algorithm. The range of individual AGBF was between approximately 7 dB (Brand B, flat configuration) and 16 dB (Brand E, flat configuration). This difference in ranges is also reflected in the standard deviation values shown in

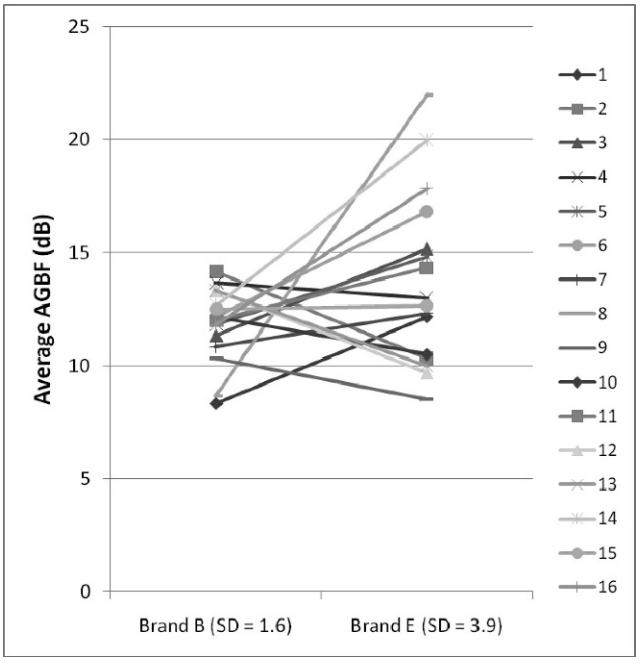


**Figure 6.** The AGBF measured in 16 participants averaged from 1000 to 3000 Hz (1/2-octave center frequencies) and across the two hearing loss configurations. Data are shown for the two FS algorithms with the largest average AGBF. The SD values represent one standard deviation for each FS algorithm.

Figures 4 and 5. Variability was comparatively small for Brands B and Cf (SD $<$ 2.6 dB) when compared to Brands D and E (SD $>$ 4.0 dB). The differences in variability are especially intriguing when considering Brands B and E. Lower variability might be expected when the AGBF is small because of possible floor effects. However, Brands B and E revealed both the largest average AGBF values and the lowest and highest variability, respectively. This finding suggests that the FS algorithm used in Brand B may have provided a more stable estimation of the feedback loop response across a range of ear shapes than Brand E. This finding has important clinical implications for those selecting hearing aids based on their expected AGBF. Specifically, while the average patient would be expected to receive approximately 12 dB of AGBF when fitted with Brand B and 14.5 dB when fitted with Brand E, individual patients might receive very different magnitudes. For Brand B, AGBF values would be expected to range from at least 7 to 16 dB (when considering a range of hearing loss configurations), while for Brand E, AGBF values would be expected to range from at least 7 to 23 dB.

The clinical significance of this finding might be further enhanced if there was an interaction between individual AGBF and the specific hearing aid. That is, while establishing that there is a range of expected AGBF in individuals is important, clinical decision making would be less affected if an individual patient's magnitude of AGBF ranked consistently across different hearing aids. However, if individual patients obtain AGBF values that are much less than the instrument average of one model and much greater than the instrument average on another model, clinical choice related to the specific model becomes much more difficult. To further examine this issue, the AGBF for Brands B and E measured in each participant were averaged across the two gain configurations and plotted in Figure 6 with a line connecting the data points for each participant. Unfortunately, these data do not support similar ranking of AGBF across two models within individuals. For example, Subject 8 obtained 22 dB of AGBF with Brand E (ranking 1st of 16), but only 8.7 dB with Brand B (ranking 15th of 16). Conversely, Subject 2 obtained 14.2 dB of AGBF with Brand B (ranking 1st of 16), but only 10.2 dB with Brand E (ranking 12th of 16). Still other participants (e.g., Subject 9) had a similar ranking across the two instruments (14th and 16th for Brands B and E, respectively).

These findings suggest that further work directed at identifying the ear specific factors important for optimal AGBF with specific FS algorithms is needed to aid in optimal clinical selection of these algorithms. An alternative argument is that the variability in AGBF observed in this study was simply due to measurement error. However, this conclusion does not appear to be supported based on at least two factors. First, the range of measured AGBF varied across the models tested and was consistently smaller for Models B and Cf than Models D and E. This finding suggests that the variability in AGBF due to measurement error was likely no larger than that found in the least variable model (Brand B, range of AGBF $<$8 dB). In addition, a large degree of measurement error was not supported by test-retest measures on the KEMAR. Specifically, the range of AGBF measured across four repetitions within the same hearing aid model was small and similar for all five models (approximately 2.5 to 3 dB). Unfortunately, test-retest reliability was not evaluated in actual patients in the current study. It is expected that the measurement range would be larger on actual patients than the KEMAR due to head movement and other factors. Therefore, it is unknown whether the 8 dB range measured for Brand B was limited by measurement error, or whether that actual measurement error was smaller than 8 dB, and Brand B also demonstrated a lack of fully robust operation across different ears (though notably more robust than models D and E).

One additional, clinically relevant difference related to the feedback produced once the hearing aid entered oscillation was also of note. In hearing aids without FS algorithms, it is common for feedback to first occur at a lower level and then increase as the gain of the instrument is increased. The feedback produced by Brands A and C (f and s) was similar to the commonly

observed behavior when the FS algorithm is absent. The initial feedback levels for Brands B and D were easily audible to the experimenter but were not extremely high level. In contrast, the feedback produced by Brand E was present and measured to be at the hearing aids OSPL90 immediately upon oscillation. Participants in this study remarked that the immediate high level feedback produced by this device was much more disturbing than that produced by the other models, though of course this feedback occurred with much higher gain settings than for Brands A and C.

## CONCLUSIONS

The results of this study are consistent with previous investigations that have shown a large range of AGBF (0–15 dB) across the FS implemented in commercial hearing aids at the time of this writing. In addition to this finding, the results also revealed considerable variability within the same FS algorithms in individual ears. This lack of robust FS across ears also varied in magnitude, as a function of the specific model with AGBF values was as small as 8 dB for Brand B and as large as 16 dB for Brand E. Unfortunately, these results suggest that on any individual patient, the largest AGBF may not be obtained from the model that reveals the largest AGBF on the average. These results also suggest that clinicians would benefit from knowing both the average and the range of expected AGBF in real ears when making selection decisions. Finally, AGBF measured in real ears and on the acoustic manikin within the same instrument suggest that across instrument comparisons of AGBF measured using acoustic manikin techniques may be misleading, especially when differences between hearing aids are small (less than 6 dB).

## REFERENCES

Barlow NL, Auslander MC, Rines D, Stelmachowicz PG. (1988) Probe-tube microphone measures in hearing-impaired children and adults. *Ear Hear* 9(5):243–247.

Boukis C, Mandic DP, Constantinides AG. (2007) Toward bias minimization in acoustic feedback cancellation systems. *J Acoust Soc Am* 121(3):1529–1537.

Dillon H. (1999) NAL-NL1: a new prescriptive fitting procedure for non-linear hearing aids. *Hear J* 52:10–17.

Dillon H, Murray N. (1987) Accuracy of twelve methods for estimating the real ear gain of hearing aids. *Ear Hear* 8(1):2–11.

Egolf DP, Haley BT, Howell HC, Legowski S, Larson VD. (1989) Simulating the open-loop transfer function as a means for understanding acoustic feedback in hearing aids. *J Acoust Soc Am* 85(1):454–467.

Freed DJ, Soli SD. (2006) An objective procedure for evaluation of adaptive antifeedback algorithms in hearing aids. *Ear Hear* 27(4):382–398.

Greenberg JE, Zurek PM, Brantley M. (2000) Evaluation of feedback reduction algorithms for hearing aid. *J Acoust Soc Am* 108(5):2366–2376.

Hellgren J, Lunner T, Arlinger S. (1999) System identification of feedback in hearing aids. *J Acoust Soc Am* 105(6):3481–3496.

Ji YS, Jung SY, Kwon SY, Kim IY, Kim S, Lee SM. (2005) An efficient adaptive feedback cancellation using by independent component analysis for hearing aids. *Conf Proc IEEE Eng Med Biol Soc* 3:2711–2713.

Johnson EE. (2006) Segmenting dispensers: factors in the selection of open-canal fittings. *Hear J* 59(11):58–64.

Johnson EE, Ricketts TA, Hornsby BWY. (2007) The effect of digital phase cancellation feedback reduction systems on amplified sound quality. *J Am Acad Audiol* 18(5):404–416.

Joson HAL, Asano F, Suzuki Y, Sone T. (1993) Adaptive feedback cancellation with frequency compression for hearing aids. *J Acoust Soc Am* 94:3248–3254.

Kaelin A, Lindgren A, Wyrsch S. (1998) A digital frequency domain implementation of a very high gain hearing aid with compensation for recruitment of loudness and acoustic echo cancellation. *Signal Process* 64:71–85.

Kates JM. (1988) A computer simulation of hearing aid response and the effects of ear canal size. *J Acoust Soc Am* 83(5):1952–1963.

Kates JM. (1999) Constrained adaptation for feedback cancellation in hearing aids. *J Acoust Soc Am* 106(2):1010–1019.

Killion MC, Revit LJ. (1987) Insertion gain repeatability versus loudspeaker location: you want me to put my loudspeaker where? *Ear Hear* 8(5 Suppl.):68S–73S.

Kochkin S. (2003) On the issue of value: hearing aid benefit, price, satisfaction, and brand repurchase rates. *Hear Rev* 10(2):12–25.

Kochkin S. (2005) Customer satisfaction with hearing aids in the digital age. *Hear J* 58(9):30–37.

Kuk F. (1994) Maximum usable real-ear insertion gain with ten earmold designs. *J Am Acad Audiol* 5:44–51.

Lantz J, Jensen OD, Haastrup A, Olsen SØ. (2007) Real-ear measurement verification for open, non-occluding hearing instruments. *Int J Audiol* 46(1):11–16.

Lee S, Kim IY, Park YC. (2007) Approximated affine projection algorithm for feedback cancellation in hearing aids. *Comput Methods Programs Biomed* 87(3):254–261.

Merks I, Banerjee S, Trine, T. (2006) Assessing the effectiveness of feedback cancellers in hearing aids. *Hear Rev* 13(4):53–54, 57.

Mueller HG, Ricketts TA. (2006) Open-canal fittings: ten take-home tips. *Hear J* 59(11):24–39.

Parsa V. (2006) Acoustic feedback and its reduction through digital signal processing. *Hear J* 59(11):16–23.

Rafaely B, Roccasalva-Firenze M, Payne E. (2000) Feedback path variability modeling for robust hearing aids. *J Acoust Soc Am* 107:2665–2673.

Ringdahl A, Leijon A. (1984) The reliability of insertion gain measurements using probe microphones in the ear canal. *Scand Audiol* 13(3):173–178.

Shin M, Wang S, Bentler R, He S. (2007). New feedback detection method for performance evaluation of hearing aids. *Sound Vibration* 302:350–360.