

Measurement of hearing aid internal noise^{a)}

James D. Lewis,^{b)} Shawn S. Goodman, and Ruth A. Bentler

Department of Communication Sciences and Disorders, University of Iowa, Iowa City, Iowa 52242

(Received 14 October 2009; revised 25 January 2010; accepted 27 January 2010)

Hearing aid equivalent input noise (EIN) measures assume the primary source of internal noise to be located prior to amplification and to be constant regardless of input level. EIN will underestimate internal noise in the case that noise is generated following amplification. The present study investigated the internal noise levels of six hearing aids (HAs). Concurrent with HA processing of a speech-like stimulus with both adaptive features (acoustic feedback cancellation, digital noise reduction, microphone directionality) enabled and disabled, internal noise was quantified for various stimulus levels as the variance across repeated trials. Changes in noise level as a function of stimulus level demonstrated that (1) generation of internal noise is not isolated to the microphone, (2) noise may be dependent on input level, and (3) certain adaptive features may contribute to internal noise. Quantifying internal noise as the variance of the output measures allows for noise to be measured under real-world processing conditions, accounts for all sources of noise, and is predictive of internal noise audibility.

© 2010 Acoustical Society of America. [DOI: 10.1121/1.3327808]

PACS number(s): 43.66.Ts, 43.50.Yw, 43.58.Ry [BLM]

Pages: 2521–2528

I. INTRODUCTION

The continual improvement of digital technology has made possible the implementation of many innovative signal processing algorithms in hearing aids (HAs). Many of these algorithms are adaptive, altering their processing based on an ongoing analysis of the temporal and spectral properties of incoming sound. With these new processing schemes, additional methods of quantifying hearing aid performance may be necessary, including the measurement of internal noise, which is currently defined as equivalent input noise (American National Standards Institute, 2004).

Internal noise is not present in the acoustic input signal, but is added by the HA itself as an unintentional by-product of its design and/or processing. Internal noise is added to any external signal processed by the HA, potentially degrading the fidelity of the sound. Internal noise can arise anywhere along the HA processing path, including at the microphone, the analog-to-digital converter (ADC), the processing algorithms, the digital-to-analog converter (DAC), and the receiver. As used here, the term “processing algorithms” refers to the application of frequency- and level-dependent gain, as well as adaptive features, such as digital noise reduction (DNR), microphone directionality, and acoustic feedback control. It has been suggested that the primary source of internal noise is the microphone, specifically, random motion of electrons in the conductors, air molecules, and the particles in the diaphragm (Thompson *et al.*, 2002). It is not unreasonable, however, to consider other components of the HA as significant contributors to internal noise as well (Stuart, 1994).

Figure 1 shows a simple model of HA internal noise. In this model, it is assumed that environmental noise is negligible, so that the only input to the HA is an acoustic signal (s). Additive internal noise can occur both before (n_1) and after (n_2) the application of gain (G). The output (L_{out}) of the HA is

$$L_{\text{out}} = G(s + n_1) + n_2. \quad (1)$$

In the United States, the current standard for quantifying HA internal noise is equivalent input noise (EIN), defined as

$$\text{EIN} = L_0 - \text{HFAG50}, \quad (2)$$

where L_0 is the level of the HA output (in dB) with no acoustic input signal present, and HFAG50 is the average gain (in dB) at 1.0, 1.6, and 2.5 kHz when the input is a 50 dB sound pressure level (SPL) pure-tone (American National Standards Institute, 2004). EIN is measured with the hearing aid at the reference test gain setting (to ensure saturation of the amplifier/compressor does not occur) and with all features (e.g., noise reduction, microphone directionality, feedback control) disabled. EIN is typically measured in a 2 cm³ coupler seated in a test box using a swept pure-tone stimulus.

An important assumption of the EIN measurement is that the primary source of internal noise is located prior to amplification (e.g., at the microphone), and any internal noise added after amplification is negligible. If this is the case, the n_2 term in Eq. (1) equals to zero. Since EIN is measured with no acoustic input signal (L_0), Eq. (1) reduces to

$$L_{\text{out}} = Gn_1. \quad (3)$$

Removal of the gain is then appropriate because it allows the internal noise to be evaluated independently of the gain of the HA. As noted earlier, however, other components of the HA may contribute significant amounts of noise, in which

^{a)} Portions of this work were presented at the annual convention of the American Acoustical Society, Scottsdale, Arizona, March 2008.

^{b)} Author to whom correspondence should be addressed. Electronic mail: james-lewis@uiowa.edu

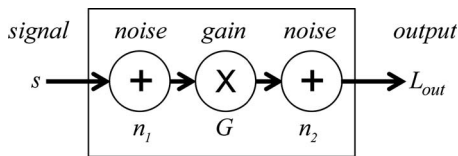


FIG. 1. Model of HA internal noise. To an acoustic signal s at the input of the hearing aid, noise (n_1) is added prior to the application of gain (G), and noise (n_2) is also added after gain.

case EIN will underestimate the internal noise of the HA. In terms of the noise model in Fig. 1, the equivalent input noise measure is arrived at by dividing both sides of Eq. (1) by the gain of the HA, resulting in

$$\text{EIN} = n_1 + \frac{n_2}{G}. \quad (4)$$

Thus, when calculating EIN, noise added after amplification (n_2) is reduced by the gain term, resulting in an underestimation of the internal noise level. From a quality control standpoint, this implies that EIN primarily evaluates microphone noise, and not noise from later-occurring sources.

The sole utility of EIN in quantifying internal noise is not only questionable from a quality control standpoint, but also in clinical utility, as EIN was not intended to measure the audibility of HA internal noise. The audibility of a HA's internal noise is dependent on the circuit noise amplitude spectrum, low-level gain, and the listener's auditory thresholds (Killion, 1976; Agnew, 1997; Thompson, 2003). The loudness or annoyance of such noise may be dependent upon the spectral shape of the noise and, thus, cannot be described by a single value (Hellman and Zwicker, 1987). Relatively limited attention in the literature is directed toward the audibility and annoyance of HA internal noise; however, investigations have typically employed 1/3 octave-band analysis of generated noise to simulate the internal noise produced by a hearing aid.

Agnew (1997) demonstrated that hearing aid internal noise becomes audible to the listener when the 1/3 octave-band level of the noise exceeds the hearing threshold at the corresponding 1/3 octave-band frequency. Furthermore, noise detection was generally the result of 500 Hz energy, presumably due to preserved hearing sensitivity, and became objectionable to some listeners at only 4 dB above threshold. Significantly, the EIN does not include 500 Hz noise in its calculation, but is limited to 1.0, 1.6, and 2.5 kHz.

Lee and Geddes (1998) investigated both audible and objectionable levels of noise among normal and hearing-impaired listeners, and concluded that designing microphones with noise levels at or below the "just-objectionable level" of the normal hearing group may be excessive, since this level is probably not audible, and certainly not objectionable, to hearing-impaired individuals. However, depending on the frequencies with the highest noise energy, individuals with certain hearing losses may be more susceptible to the audibility of internal noise than others, specifically, those with better low- and mid-frequency hearing sensitivity.

Other research suggests that the amount of background noise an individual can tolerate or judges as acceptable, de-

defined as the acceptable noise level (ANL), may be an important predictor of successful hearing aid use (Nabelek *et al.*, 1991, 2006). Additionally, relatively small changes in the ANL can have significant effects on the probability for success with hearing aids. Nabelek *et al.* (2006) found that if a subject's ANL decreases by 4 dB, the probability for success with hearing aids can increase from 45% to 92% (for an individual with an initial ANL of 10 dB). The presence of HA internal noise may also be a contributing factor to lack of success with HAs, especially if judged to be at unacceptable levels.

If a clinician encounters a patient with complaints about either background noise or the noise produced by their hearing aid, the complaint may be difficult to evaluate because a method for quantifying internal noise in a way that has perceptual relevance does not exist. Using EIN to confirm these complaints presents three main difficulties: (1) Even if noise is present at the microphone only, EIN quantifies the noise as an average at only three frequencies and does not take into account the whole audiogram; (2) if noise is not isolated to the microphone, any noise that might be present after gain is applied is not accurately evaluated by EIN; and (3) the internal noise in real-world processing conditions is not evaluated by EIN. Without other resources for measuring noise, some clinicians might be tempted to compare HAs from different manufacturers, based on EIN measures; however, such a decision may not be justified or be expected to result in a HA with the least audible internal noise.

Finally, clinicians typically use the EIN measure as a diagnostic indicator of HA performance: EIN values exceeding tolerance suggest that the aid is "out of specs" and may need repair. In-house measures according to American National Standards Institute (2004) suggest considerable variability in the EIN measure (Fig. 2). This variability was apparent across three different measurement systems, including a custom laboratory set-up in a sound-treated booth, a Frye Electronics Fonix system, and an AudioScan Verifit system. The EIN measure will erroneously label a HA as malfunctioning nearly one-third of the time.

The current study proposes and evaluates a novel method of measuring the internal noise of HAs that may be predictive of audibility to the user. The new measure may also have application as a quality control standard. The measurement quantifies the internal noise as the variance in the HA output across multiple measurement trials for a calibrated speech-like test signal. The measurement is made during hearing aid processing of the test signal, and is expected to be sensitive to all potential sources of internal noise. Internal noise is assessed across a broad range of frequencies and stimulus intensity levels, which may be beneficial when predicting noise audibility.

II. METHODS

A. Stimuli

The International Speech Test Signal (ISTS; IEC, 2008) was used as the stimulus. This 60-s recording is composed of speech segments from six different languages concatenated to yield a signal with temporal and frequency characteristics

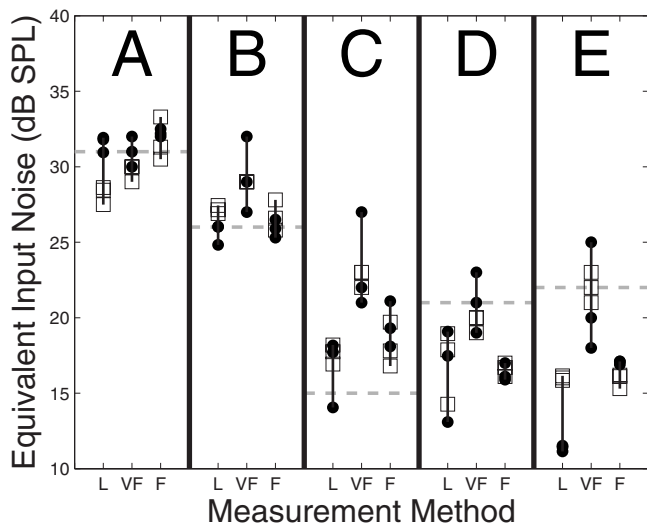


FIG. 2. Variability in clinical EIN measurements. Each panel (A–E) indicates HA model from one of six different manufacturers. EIN values are shown for two HAs of each model (filled circles, open squares). EIN values were computed using three different measurement systems (L—custom laboratory set-up, VF—AudioScan Verifit, F—Fonix 6500). Measurements were repeated three times with complete setup and takedown for each measurement. Dashed gray lines show manufacturer-reported EIN values plus 3 dB (ANSI tolerance). All HAs exceeded the ANSI tolerance of 3 dB above the manufacturer specified EIN level on at least one measurement.

representative of speech in numerous languages. ISTS was selected as a stimulus instead of the more traditional ICRA speech noise because the temporal structure of normal speech is preserved in the ISTS stimulus, decreasing the likelihood that certain noise canceling features of HAs would be activated during testing, thereby treating the signal as if it were unwanted noise.

B. Measurement paradigm

Stimulus delivery and response acquisition were controlled using a personal computer, custom software, and a sound card (Lynx L22, 24-bits, 44.1 kHz sample rate). The output of the sound card was routed through a power amplifier (ADCOM GFA 5002) to a loudspeaker (Tonoy System 800) positioned 1 m from a Knowles Electronics Manikin for Acoustic Research (KEMAR) at 0° azimuth relative to the geometric center of the head. KEMAR, as opposed to a 2 cm^3 coupler, was used to allow for more accurate predictions of internal noise audibility, as measured levels would be more representative of those encountered in the human ear canal. Hearing aid output was measured with a G.R.A.S. IEC-711 ear simulator attached to the right ear of the manikin. Multiple measurements were taken, with recording of the hearing aid output time-locked with acoustic stimulus presentation.

Responses were collected from HAs from each of the six major manufacturers. For each manufacturer, the HA model with the most advanced signal processing algorithms was selected for evaluation. The hearing aids were all set to default manufacturer (“first-fit”) settings for a user with a sloping high-frequency hearing loss (thresholds were 10, 10, 30, 45, 60, and 75 dB HL at 0.25, 0.5, 1, 2, 4, and 8 kHz, respectively). Measurements were made in a sound booth,

with additional acoustic treatment to reduce echoes and standing waves. All features (e.g., acoustic feedback control, noise reduction, directionality) were enabled or disabled as a parameter of testing.

Prior to recording the HA output for analysis, the 60-s ISTS noise was presented continuously for 8 min. This allowed the HA time to “settle,” i.e., to reach a point where it would produce the same output for a given speech input. In practice, this length of time was never needed for a HA to reach steady state (longest required was 3 min); the 8-min length was simply chosen as a very conservative value. Immediately following the settling period, a 1.4-s segment of the 60-s ISTS noise sample was selected and consecutively presented 32 times. The same segment was used for all measurements and allowed for numerous recordings to be made within a relatively short period of time. The procedures just described were repeated for two HA conditions (adaptive features enabled and disabled), at seven stimulus input levels (20–80 dB SPL in 10-dB steps), and once with no stimulus. The order of presentation was always silence first, then stimulus level from lowest to highest. All measurements were completed on two hearing aids of each model to determine measurement reliability.

C. Calculation of internal noise

Data were analyzed using custom MATLAB software. For each condition, $K=32$ independent buffers of data were collected (where one buffer represents the waveform recorded in response to a single presentation of the speech-1.4 s stimulus). Each buffer was of length $N=61\,740$ samples (1.4 s at 44.1 kHz sampling rate). The data were filtered using a high-pass finite impulse response (FIR) filter (354 Hz cutoff, 5.7 ms group delay).

Discrete Fourier transforms were computed on each buffer,

$$X_k[m] = \sum_{n=0}^{N-1} x_k[n] e^{-j2\pi mn/N}, \quad (5)$$

where x_k is the waveform in the k -th buffer, and $X_k[m]$ is the DFT of the k -th buffer and the m -th frequency bin. In the frequency domain, the signal was taken as the (coherent) average complex spectrum,

$$\bar{X}[m] = \frac{1}{K} \sum_{k=1}^K X_k[m]. \quad (6)$$

The energy in the (mean) signal is therefore

$$|\bar{X}[m]|^2 = \frac{1}{K^2} \left| \sum_{k=1}^K X_k[m] \right|^2. \quad (7)$$

Noise was defined as any part of the measured waveform which was not repeatable (i.e., it did not phase lock to the stimulus). By this definition, the energy in the hearing aid noise floor is equivalent to the variance of the measurements,

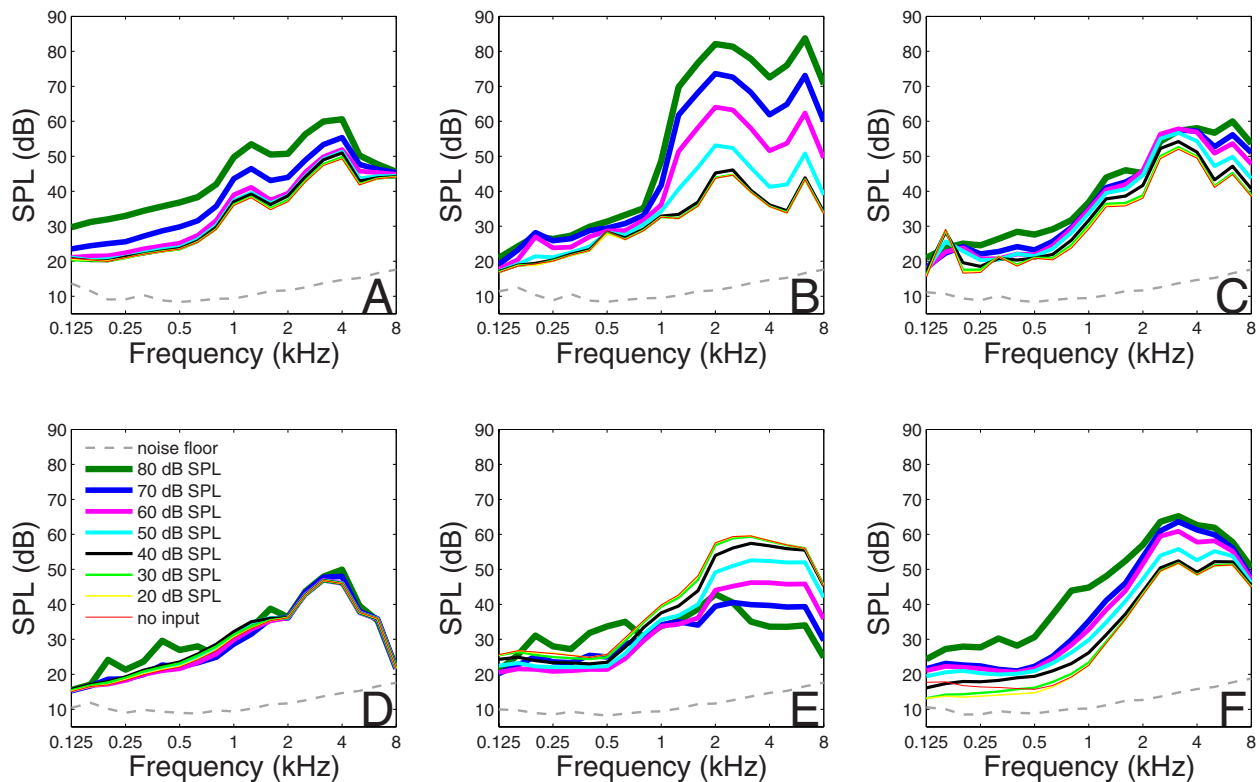


FIG. 3. (Color online) Measured internal noise (variance) as a function of frequency at various input levels for hearing aids A–F. Measurements show the internal noise generated by each hearing aid for inputs ranging from no-input (thinnest lines) to 80 dB SPL (thickest lines). The noise floor of the measurement system is plotted as the thin broken gray line.

$$S^2[m] = \frac{1}{K-1} \sum_{k=1}^K (X_k[m] - \bar{X}[m])(X_k[m] - \bar{X}[m])^*, \quad (8)$$

where $*$ denotes the complex conjugate. Defining the reference pressure as $P_{\text{ref}}=20 \mu\text{Pa}$, the spectral sound pressure levels of the signal and noise defined from frequency bins $m=1$ up to $N/2-1$ are

$$\text{SPL}_{\text{signal}} = 10 \log_{10} \left(\frac{2 |\bar{X}[m]|^2}{N^2 P_{\text{ref}}^2} \right) \quad (9)$$

and

$$\text{SPL}_{\text{noise}} = 10 \log_{10} \left(\frac{2 S^2[m]}{N^2 P_{\text{ref}}^2} \right). \quad (10)$$

An advantage of defining noise in this way is that the noise floor of a hearing aid can be measured while the aid is processing a signal and providing varying amounts of gain, for example, under compression or expansion schemes. Further, harmonic distortion or distortion products produced by the hearing aid are not calculated as noise, so long as they are exactly reproducible across repeated presentations of the same stimulus.

One potential disadvantage of defining noise in this way is the inability to distinguish between variability arising from intrinsic circuit noise versus variability in a hearing aid's amplification strategy. That is, if a hearing aid applies an algorithm that changes its response over time, the varying output will show up as increased variance, and therefore, "noise," even though this variability is too slow to be per-

ceived by human users as noise. This potential problem can be overcome if some metric of output level is plotted over time. Should the hearing aid alter its processing, this will show up as a slow drift in the output level over time, and the measurement paradigm can be altered to avoid this confound.

III. RESULTS

A. Adaptive features enabled

Internal noise as a function of frequency, with adaptive features enabled, is plotted in Fig. 3 for hearing aids A–F. The lines indicate measurements made at different stimulus levels. In all measurements, the internal noise levels of the aids were greater than the noise floor of the system and testing environment (broken, light gray line). Three basic patterns of noise can be seen in the data: (1) The internal noise increases as the stimulus level increases (Fig. 3, A, B, C, and F); (2) the internal noise is constant across stimulus levels (Fig. 3, D); and (3) the internal noise decreases as stimulus level increases (Fig. 3, E).

Based on the noise model in Fig. 1, two general patterns of noise growth were expected. When the dominant source of internal noise is located prior to gain [$n_1 \gg n_2$ in Eq. (1)], the measured noise will vary with input level. Assuming the gain decreases as input level increases, and that the internal noise is constant, the noise measured at the HA output will also decrease as level increases. When the HA has significant levels of constant internal noise both before and after the gain [$n_1 = n_2$ in Eq. (1)], the same pattern is seen. The mea-

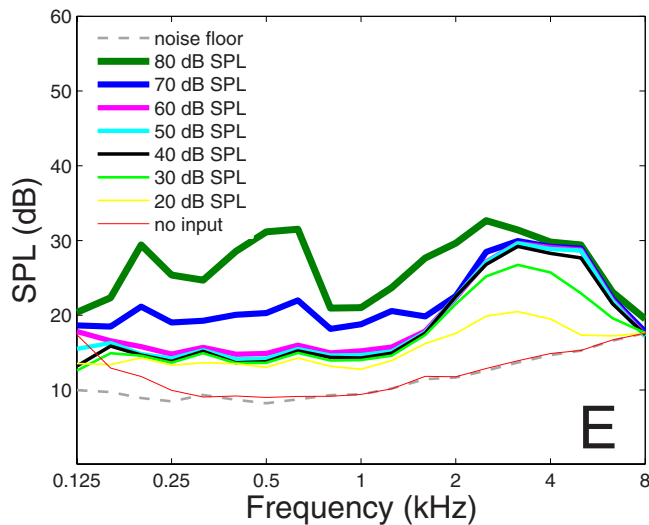


FIG. 4. (Color online) Internal noise as a function of input level for HA-E after removal of gain. Line thickness indicates level from lowest level (thinnest) to highest (thickest). The noise floor of the measurements system is plotted as the thin broken gray line.

sured noise will decrease with level, but the noise will be shifted upward by a constant amount, n_2 . Of the six HAs tested, only HA-E (Fig. 3, E) showed this pattern. It may be possible to distinguish between $n_1 \gg n_2$ and $n_1 = n_2$ by removing the gain from the noise measurements. When this is done, a system characterized by $n_1 \gg n_2$ will show noise that is constant with level, while a system characterized by $n_1 = n_2$ will show noise that varies with level. Figure 4 shows the internal noise of HA-E with the gain removed. The noise with the gain removed is fairly constant through the mid-range of input levels, but not at the lower and higher levels. These results are most consistent with noise being present both before and after amplification.

The second pattern of expected noise growth occurs when the dominant source of internal noise is located after the gain [$n_1 \ll n_2$ in Eq. (1)]. In this case, the measured noise will be constant with input level, again assuming that the internal noise is constant. Hearing aid D (Fig. 3, D) showed this pattern of noise. At all but the highest two levels, HA-A (Fig. 3, A) also showed this pattern.

The remaining three HAs (B, C, and F) showed an increase in internal noise as input level increased (and as gain decreased). The cause of this pattern is unclear. One possibility is that the internal noise is not constant as a function of level. In particular, it appears that the internal noise increases as input level increases. As noted earlier, this increase noise was measured as an increase in the variance of the HA output, so that this increase is not expected to reflect an increase in harmonic or other distortion at high levels. It is possible that the patterns of noise growth are caused by a complex interaction between noise occurring before gain (n_1) and noise occurring after gain (n_2).

Figure 5 shows the change in internal noise level across two HAs of the same model (HA-E). Measured noise levels varied by approximately 2 dB across this HA model with the exception of the most intense input level where a difference

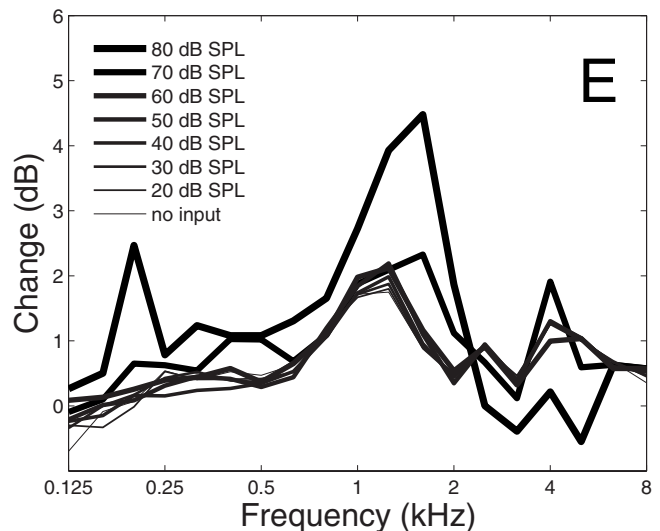


FIG. 5. Difference in internal noise levels across two hearing aids of the same model (HA-E) when tested individually and programmed for the same hearing loss. Line thickness indicates level from lowest level (thinnest) to highest (thickest).

of nearly 5 dB was observed. These results were typical of the reliability across hearing aids of the same model for the remainder of the groups.

B. Adaptive features disabled

Figure 6 shows the change in internal noise levels when adaptive features were disabled, relative to adaptive features enabled. Each panel shows a different frequency. Results suggested that enabling features such as feedback cancellation, directionality, and digital noise reduction made only a small contribution to the internal noise of the majority of the HAs (as tested in a quiet room with a single sound source).

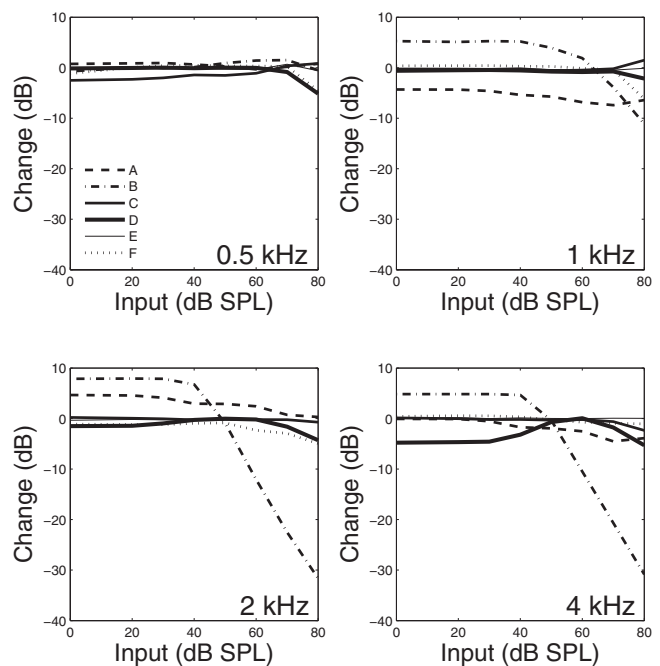


FIG. 6. Change in variance when adaptive features are disabled as a function of input level. Each panel represents a different frequency. Each line in a panel represents a different HA.

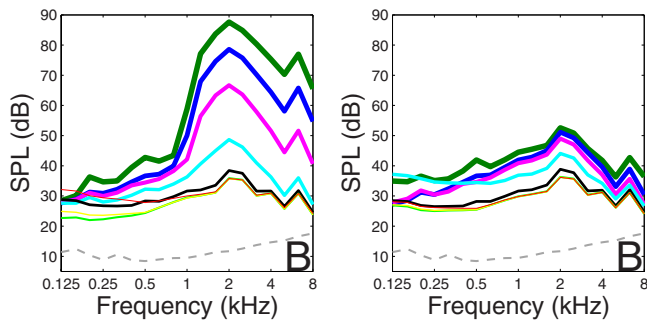


FIG. 7. (Color online) Internal noise of HA-B with adaptive features enabled (left panel) and with feedback management disabled (right panel) for each input level. Line thickness indicates level from lowest level (thinnest) to highest (thickest). The thin gray broken line indicates measurement system noise floor.

Except for HA-B, noise levels with features disabled remained within 5 dB of levels measured with features enabled. When adaptive features were enabled, the internal noise in HA-B grew nearly linearly with higher intensity levels at frequencies above 1 kHz (Fig. 3, B). When adaptive features were disabled in HA-B, noise levels were reduced by as much as 30 dB. Figure 7 shows the internal noise of HA-B with adaptive features enabled (left panel) and adaptive features disabled (right panel). With features disabled, noise levels were similar to the other HAs tested (Fig. 3). Further testing revealed that it was specifically the acoustic feedback cancellation feature that was the cause of this noise. This particular HA used a phase modulation algorithm to control feedback. Because the variance method of measuring

noise is phase sensitive, it is probable that activation of the algorithm during the measurement process led to an increase in the measured noise.

C. Audibility of internal noise

The noise levels of HA-B generated for the 80 dB SPL input approach nearly 90 dB SPL when quantified by the variance in the measurements; however, listening tests carried out by the authors suggested that such levels are not perceived by human listeners. The discrepancy between perceived and measured noise appears to be due to the phase sensitivity of the variance method for calculating noise and the general phase insensitivity of the human auditory system. Further research is needed regarding users perception of such phase-altering algorithms and their relationship to noise measurements.

The most likely condition in which hearing aid users will hear the internal noise of the HA is a quiet environment. The HA internal noise levels, measured in a sound-treated booth, ranged from 15–60 dB SPL across the frequency range of 0.125–8 kHz (Fig. 3). Figure 8 compares the noise levels in quiet and the audiometric thresholds of the hearing loss for which the HAs were programmed. Wherever the thresholds (solid lines) fall below the noise level (broken lines), the noise was predicted to be audible to a listener having the same audiometric thresholds. The internal noise levels of all aids were predicted to be audible, with the exception of HA-F. In all cases where the noise was audible, noise levels exceeded hearing thresholds in the frequency range of 0.5–1.5 kHz.

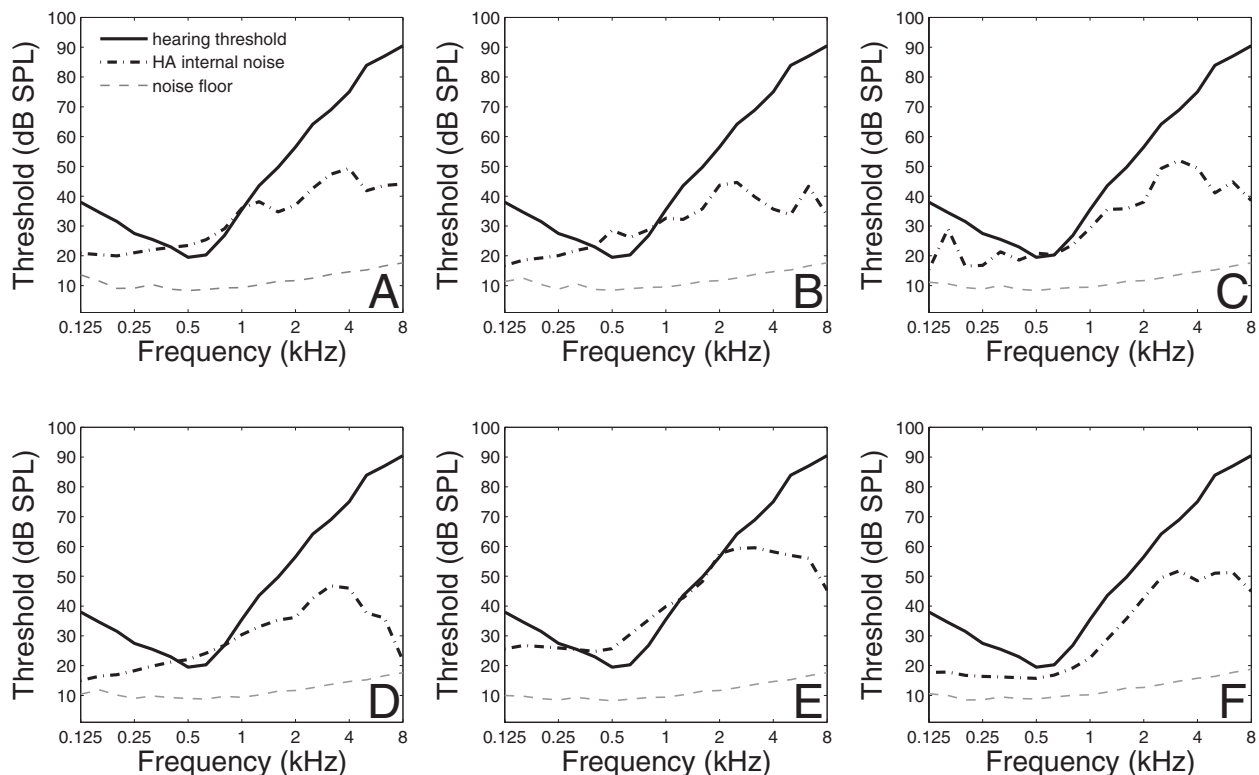


FIG. 8. Internal noise in quiet (dark broken lines) and programmed audiometric thresholds (solid lines) for each HA. Internal noise is predicted to be audible where the noise level is equal to or greater than the associated hearing threshold.

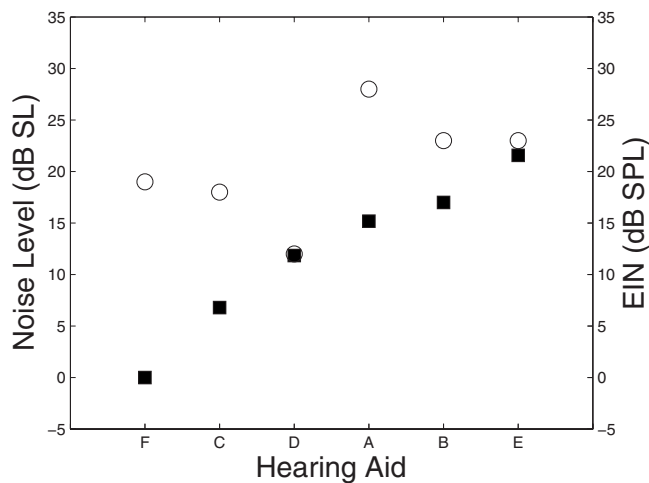


FIG. 9. Total audibility levels (filled squares) and manufacturer specified EIN values (open circles) for each hearing aid.

The “total audibility” for each aid was determined by subtracting the hearing threshold (dB SPL) from the noise level (dB SPL) at each 1/3 octave frequency where the noise was predicted to be audible. The differences at each frequency were converted into linear units, summed and converted back into decibels. Figure 9 compares the total audibility levels (filled squares) of the six HAs (arranged in order of increasing total audibility) and the manufacturers’ reported EIN levels (open circles). Since the noise of HA-F was not predicted to be audible, a level of 0 dB SPL was assigned to it for display purposes only. This figure demonstrates that, although a particular HA may have a very low EIN value as quantified by American National Standards Institute (2004), the actual audibility of the internal noise to the hearing-impaired user may be higher than similar HAs with higher EIN values. For example, HA-D had the lowest reported EIN value (12 dB SPL), but two other HAs (F and C) were predicted to have internal noise levels that are less audible. As expected, the results reinforce the fact that EIN was never meant to indicate audibility. Further testing should be done to see if the relative audibility predictions based stemming from the variance measurements of internal noise are psychophysically valid.

IV. DISCUSSION

The current EIN calculation assumes a model of internal noise where the dominant noise source is located prior to the application of gain. When this is the case, internal noise measured at the output of the receiver and quantified as the variance across repeated trials is expected to change in proportion to the gain. As the input level of the stimulus increases, less gain will be applied and the noise will decrease accordingly. This pattern of noise was only apparent for one of the six hearing aids tested (HA-E). A model in which internal noise is primarily located after amplification is consistent with the measurements for HA-D, while a model with internal noise sources located both before and after gain is possible for the remaining four HAs.

Noise generated prior to amplification logically stems from the microphone (Thompson *et al.*, 2002); however,

noise generated following amplification may result from several different mechanisms. One possible source of internal noise is the processing algorithms employed by the hearing aid, including acoustic feedback management, microphone directionality, and DNR. All six HAs used in this study included these specific features and are representative of nearly all modern hearing aids today. With one exception (HA-B), the results of this study suggested that these features do not significantly contribute to the internal noise of a HA. However, the testing environment used in the current study should be considered. All measurements were made in a quiet, sound-attenuating booth using a single sound source. Features such as directionality and digital noise reduction are designed to only be active in noisy environments, and may therefore not have been engaged by the processing of the hearing aid. Depending on the coupling between the hearing aid and KEMAR, feedback management may or may not have been engaged throughout testing.

The observation of internal noise changing due to activation of adaptive features for HA-B demonstrates both a strength and weakness of the proposed method to measure internal noise. It is apparent from these results that certain processing algorithms do contribute to the HA internal noise, and may or may not reduce the fidelity of the signal presented to the user. This is not something that can be determined by the current EIN calculation since it is calculated following measurements of hearing aid output in the absence of a complex input. The advantage then of quantifying the internal noise of the aid as the variance across multiple stimulus presentations is that the noise present in the processed complex signal is known.

Accordingly, the noise levels measured in terms of output variance provide insight into the auditory perception of the noise experienced by a HA user under normal-use conditions. Each hearing aid, except HA-F, was predicted to have noise levels audible to an individual with a high-frequency sensorineural hearing loss. The frequency distribution of the internal noise was similar for all the aids, and whenever noise was predicted to be audible, it was generally associated with energy around 750 Hz, due to the preservation of hearing sensitivity. The broad range of “total audibility” levels across the six HAs implies that the internal noise levels of some HAs may be significantly less than that of other HAs when programmed for an identical loss. For example, the total audibility for HA-E was predicted to be nearly 20 dB greater than the value calculated for HA-C (Fig. 9). In light of the research suggesting HA user intolerance to noise levels only slightly above threshold (Agnew, 1997), and the impact of background noise on successful hearing aid use (Nabelek *et al.*, 2006), quantifying noise as the output variance could be beneficial to clinicians in selecting a “quiet” hearing aid for a patient who is either a poor HA user or who is expected to be a poor user. As demonstrated in Sec. III, the EIN calculation is not an accurate predictor of noise audibility and should not be used as such. Further research is needed to verify the audibility predictions of the internal noise measurement method proposed in this paper.

In the case of the six aids assessed, HA-F would seem to be the obvious choice for a patient who is particularly sensitive to noise with an associated high ANL. However, it should be kept in mind that the measurements reported in the present study were obtained after each aid was programmed to its manufacturer's first-fit algorithm. Each manufacturer employs different amplification strategies, and therefore each aid amplified the test stimuli using different amounts of gain. From a quality control standpoint, to accurately compare aids, it would first be necessary to ensure all aids are providing equal gain. From a clinical standpoint, however, first-fit algorithms are commonly used, and gain differences often exist across aids despite similar audiograms, unless care is taken to match output levels using real ear verification.

As noted earlier, a potential short-coming to using output variance to quantify internal noise is that phase manipulations performed by a HA will add to, and may even dominate the noise measurement. This is undesirable if one of the goals of the measure is to predict noise audibility. It seems likely that such phase inconsistencies in processing are inconsequential, although it also seems that consistent processing of identical stimuli is desirable. This is a topic for further studies.

V. CONCLUSIONS

The current study described a method of quantifying the internal noise of hearing aids as the output variance across repeated trials. Internal noise levels were measured at stimulus input levels ranging from 20–80 dB SPL, and also in the absence of an input. Patterns of internal noise level as a function of stimulus level (and therefore gain) challenge the basic assumption of the current EIN calculation that the noise floor of a hearing aid is dominated by the microphone's contribution. Of the six aids tested, only one aid (HA-E) was consistent with the EIN model of internal noise. One aid (HA-D) had noise levels consistent with a primary noise generator following amplification. The remaining aids demonstrated more complicated models of non-constant internal noise that may have been generated, both prior to and following amplification.

In contrast to the EIN, calculating internal noise as the output variance across repeated trials allows for measurements of noise concurrent with HA processing of real-world

stimuli. This may provide insight into the contribution of processing algorithms to the overall noise floor. One caveat is that phase manipulations, which contribute to the variance, may not be perceived by listeners.

Quantifying the noise as the variance permits predictions of noise audibility to be made, and may be beneficial from the standpoint of both quality control and clinical utility.

ACKNOWLEDGMENTS

The investigators thank the National Institutes of Health and the American Auditory Society (AAS) for funding the presentation of this study at the AAS 2008 Annual Convention (Mentored Doctoral Student Research Poster Session Grant). Additionally, the investigators thank the following manufacturers for providing the hearing aids used in this study: Phonak, Unitron, Widex, Starkey, Siemens, and Oticon.

- Agnew, J. (1997). "Audible circuit noise in hearing aid amplifiers," *J. Acoust. Soc. Am.* **102**, 2793–2799.
- American National Standards Institute (2004). *American National Standards Specification for Audiometers (ANSI S3.6–1996)* (American National Standards Institute, New York).
- American National Standards Institute (2004). *American National Standards Specification of Hearing Aid Characteristics (ANSI S3.22–2004)* (American National Standards Institute, New York).
- Hellman, R., and Zwicker, E. (1987). "Why can a decrease in dB (A) produce an increase in loudness?," *J. Acoust. Soc. Am.* **82**, 1700–1705.
- International Electrotechnical Commission (2008). *Electroacoustics-Hearing Aids—Part 15: Methods for characterizing signal processing in hearing aids (IEC 60118–15)* (International Electrotechnical Commission, Geneva, Switzerland).
- Killion, M. C. (1976). "Noise of ears and microphones," *J. Acoust. Soc. Am.* **59**, 424–433.
- Lee, L. W., and Geddes, E. R. (1998). "Perception of microphone noise in hearing instruments," *J. Acoust. Soc. Am.* **104**, 3364–3367.
- Nabelek, A. K., Freyaldenhoven, M., Tampas, J., and Burchfield, S. (2006). "Acceptable noise level as a predictor of hearing aid use," *J. Am. Acad. Audiol* **17**, 626–639.
- Nabelek, A. K., Tucker, F. M., and Letowski, T. R. (1991). "Toleration of background noises: Relationship with patterns of hearing aid use by elderly persons," *J. Speech Hear. Res.* **34**, 679–685.
- Stuart, J. R. (1994). "Noise: Methods for estimating detectability and threshold," *J. Audio Eng. Soc.* **42**, 124–140.
- Thompson, S. C. (2003). "Internal noise in hearing aid microphones: Its causes and consequences," *Hear. J.* **56**, 46–47.
- Thompson, S. C., LoPresti, J. L., Ring, E. M., Nepomunceno, H. G., Beard, J. J., Ballard, W. J., and Carlson, E. V. (2002). "Noise in miniature microphones," *J. Acoust. Soc. Am.* **111**, 861–866.