AND9020/D

Adaptive Feedback Cancellation 3 from ON Semiconductor

ON Semiconductor®

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APPLICATION NOTE

INTRODUCTION

This information note describes the feedback cancellation feature provided in ON Semiconductor's latest digital hearing aid amplifiers, AFC3. Feedback cancellation differs from other feedback reduction strategies in that it does not reduce the forward gain of the hearing aid under normal operating conditions. Instead, the algorithm reduces feedback by formulating an internal estimate of the true hearing aid feedback signal and then subtracting this estimate from the microphone signal. If the internal estimate is accurate, then the actual feedback is cancelled and the hearing aid will not squeal.

Maintaining an accurate estimate of the true hearing aid feedback signal is essential to proper operation of the algorithm. Since the feedback signal changes over time, the algorithm must make continuous adjustments of the internal model to ensure accurate matching of the external feedback. In ON Semiconductor's feedback canceller, these adjustments occur automatically and do not require probe signals or other user interventions that might interfere with the operation of the hearing aid.

The sections below describe feedback canceller operation in more detail highlighting factors that affect feedback canceller performance. The article concludes with a discussion of methods that can be used to optimize feedback canceller performance. These methods allow users of ON Semiconductor digital amplifiers to increase the performance of their products.

PRINCIPLE OF OPERATION

AFC3, from ON Semiconductor, is a true feedback cancellation (also known as *phase cancellation*) algorithm. The algorithm operates by subtracting an internal estimate of the hearing-aid feedback signal from the microphone signal. The internal feedback signal is obtained by passing the hearing-aid output signal through an internal, feedback-path model.

The feedback-path model in AFC3 is intended to match the impulse response, or time response, of the external feedback path. The feedback-path impulse response is the time signal that would be observed at the microphone output if an impulsive signal were produced by the hearing aid. As an example, the measured impulse response of a typical BTE hearing aid is shown in Figure 1−a. This plot shows an initial period of approximately 0.5 − 0.75 ms with little or no energy. This time window corresponds to the analog-to-digital and digital-to-analog converter delays as well as any acoustic delays due to transducers, tubing, sound propagation through the air, etc. Following the initial delay is the main part of the feedback-path response which, in this example, persists for more than 2 ms.

The net effect of the feedback canceller algorithm is to remove a certain portion of the feedback path impulse response. Since computational resources inside a hearing-aid signal processor are limited, only a portion of the total feedback path response can be cancelled. To achieve maximum effectiveness of the canceller, the cancelled part of the response should be that portion of the feedback path with the largest energy. This ensures that the net response has the lowest energy for the given length of canceller. For example, if the cancellation window is restricted to 1 ms, then for the feedback path of Figure 1−a, the internal feedback model should look like the plot of Figure 1−b. When this is subtracted from the feedback−path response, the net feedback path is as shown in Figure 1−c. The corresponding frequency response is shown in Figure 2. This figure illustrates that the net effect of the feedback canceller in this example is to reduce the peak response of the feedback path by approximately 12 dB. Consequently, the feedback canceller in this example should provide approximately 12 dB of added stable gain (ASG) that is achieved without any impact on the forward gain of the hearing aid.

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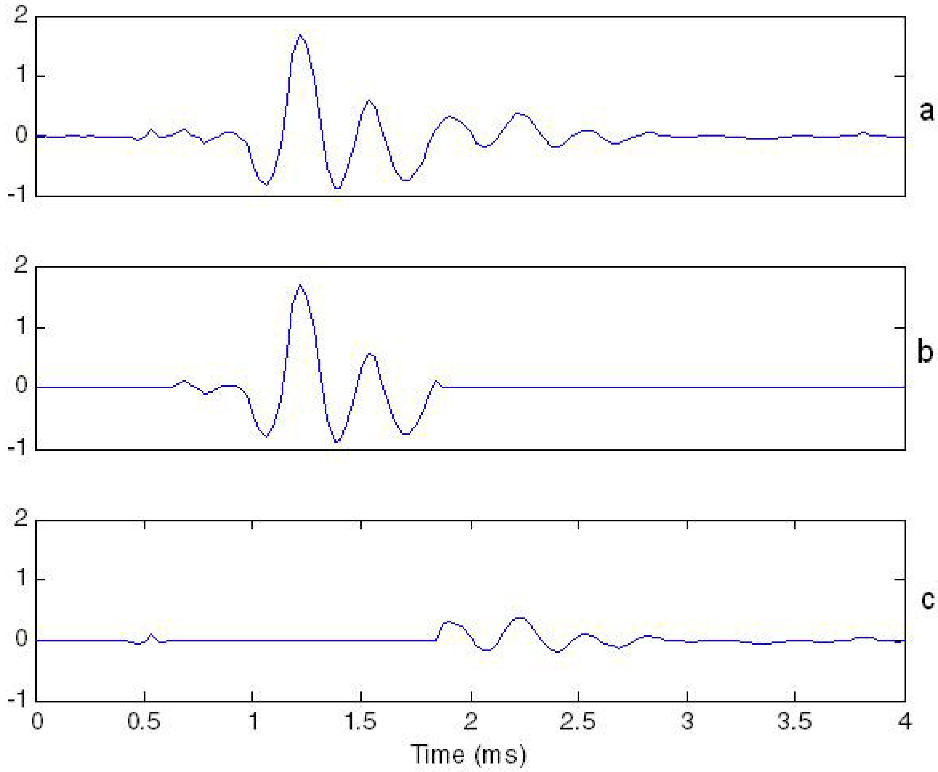

Figure 1. Measured impulse responses of a typical BTE: a) total external feedback path, b) internal feedback path of the feedback canceller, c) net effect including the effect of the feedback canceller

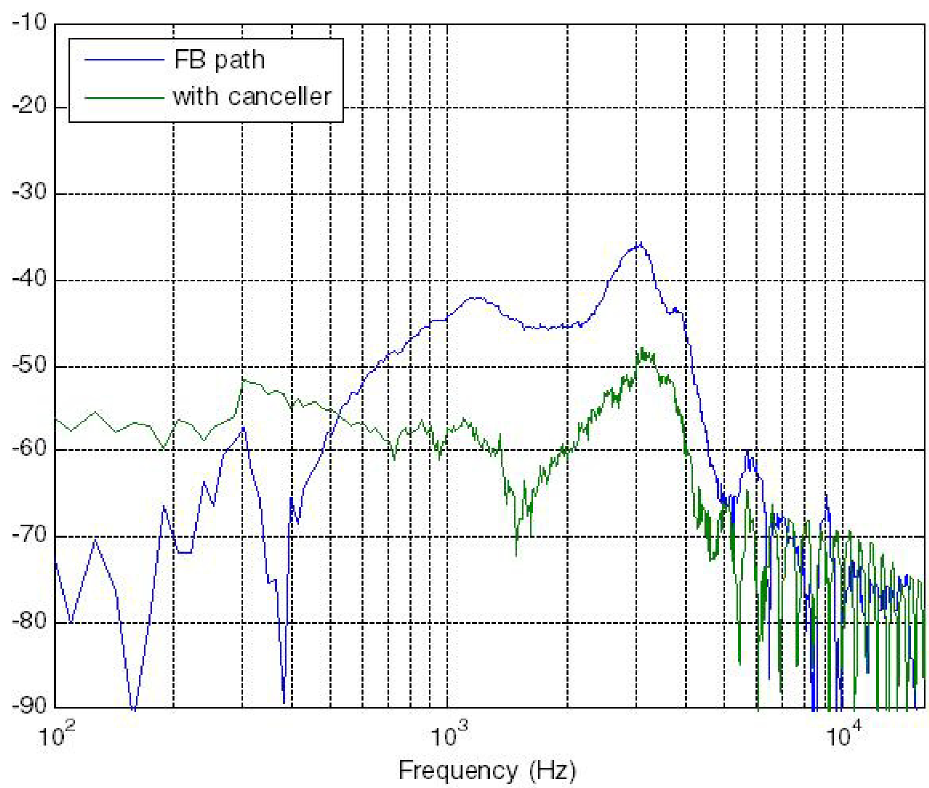

Figure 2. Change in feedback−path frequency response before and after the application of feedback canceller shown in Figure 1.

Unfortunately, the feedback path of a hearing aid is not constant. There is significant variability in the feedback paths of different hearing aids due to factors such as transducer selection, acoustic tubing, ear mold styles, etc. Furthermore, the feedback path of a specific hearing aid can also change with time due to user movements and environmental changes. If such changes are not reflected in the internal feedback path, then canceller performance can be significantly degraded.

To maintain optimal cancellation performance, AFC3 performs continuous updates of the internal feedback-path model. These updates occur automatically and utilize the existing hearing aid signals. No user intervention or special probe signals are required thus minimizing disruption for the hearing-aid wearer.

OPTIMIZING PERFORMANCE

Although conceptually simple, the performance of a feedback canceller algorithm in a hearing aid is influenced by many system-level factors. Feedback canceller performance is influenced by hearing-aid style, acoustic response, audio signal bandwidth, input-signal characteristics and interactions with other hearing-aid features. The following sections outline these influences and describe strategies for achieving optimum feedback canceller performance.

Influence of Hearing Aid Style

Hearing-aid styles influence feedback canceller performance because of the differences in feedback-path acoustic delay. Simply put, the time required for sound to propagate from the receiver back to the microphone is different for an ITC compared to a BTE. The acoustic propagation time adds to the converter delays and appears at the start of the feedback path impulse response. This delays the onset of the maximum-energy segment of the feedback-path response. To maximize canceller performance, it must be compensated by the internal model.

As an example, the impulse responses of three different hearing aids (ITC, ITE and BTE) are shown in Figure 3. Note how the time onset of the impulse response occurs later for the hearing aid styles with the larger receiver/microphone separation.

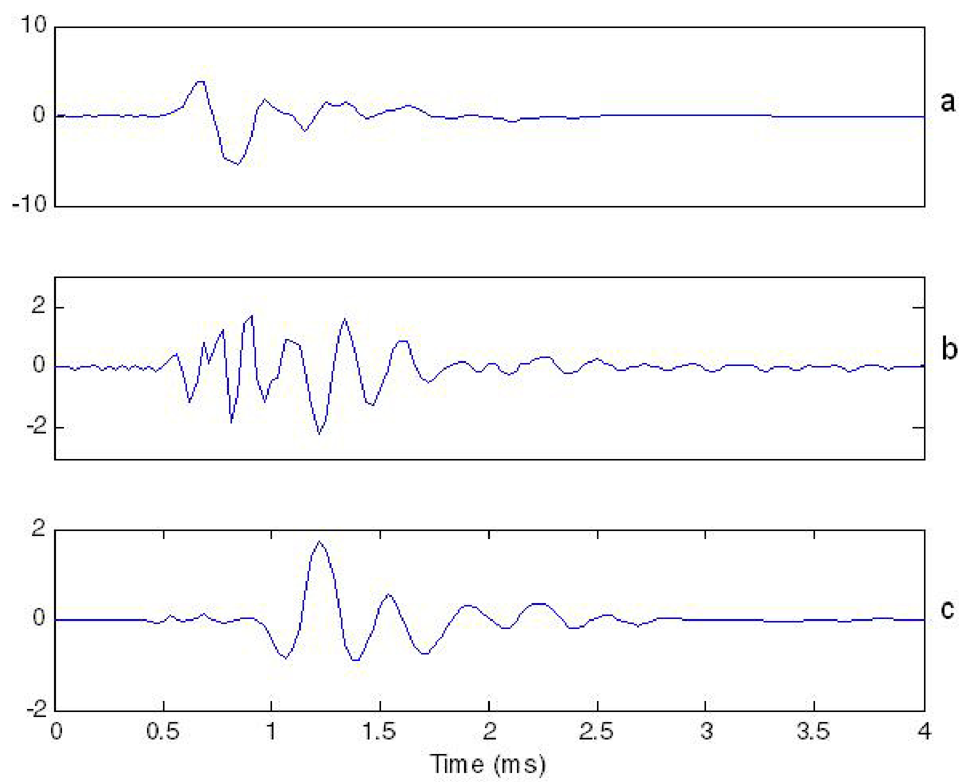

Figure 3. Impulse responses from three different hearing aid styles: a) ITC, b) ITE, and c) BTE. Notice the increased initial delay for the larger styles.

AFC3 from ON Semiconductor accommodates different hearing aid styles through an adjustable *acoustic predelay* parameter. This parameter allows the time location of the feedback canceller operation to be optimized for different hearing aid styles without adding any computational complexity or power consumption to the hearing aid circuit.

To illustrate the impact of the acoustic predelay on AFC3 performance, the added stable gains were measured for five different hearing aid styles: ITC, ITE, BTE, BTE-open and RITE (Receiver In The Ear). ASG measurements were made for many different acoustic predelay settings for each hearing aid style. The results are shown in Figure 4.

As shown in Figure 4, ASG peaks at different acoustic predelays for the different hearing-aid styles. Generally, hearing aids with a larger physical separation between microphone and receiver (BTE, BTE-O and RITE) required a longer acoustic predelay to achieve the highest ASG. Also, using an acoustic predelay that is not suited for the particular hearing-aid style can impair the achievable ASG.

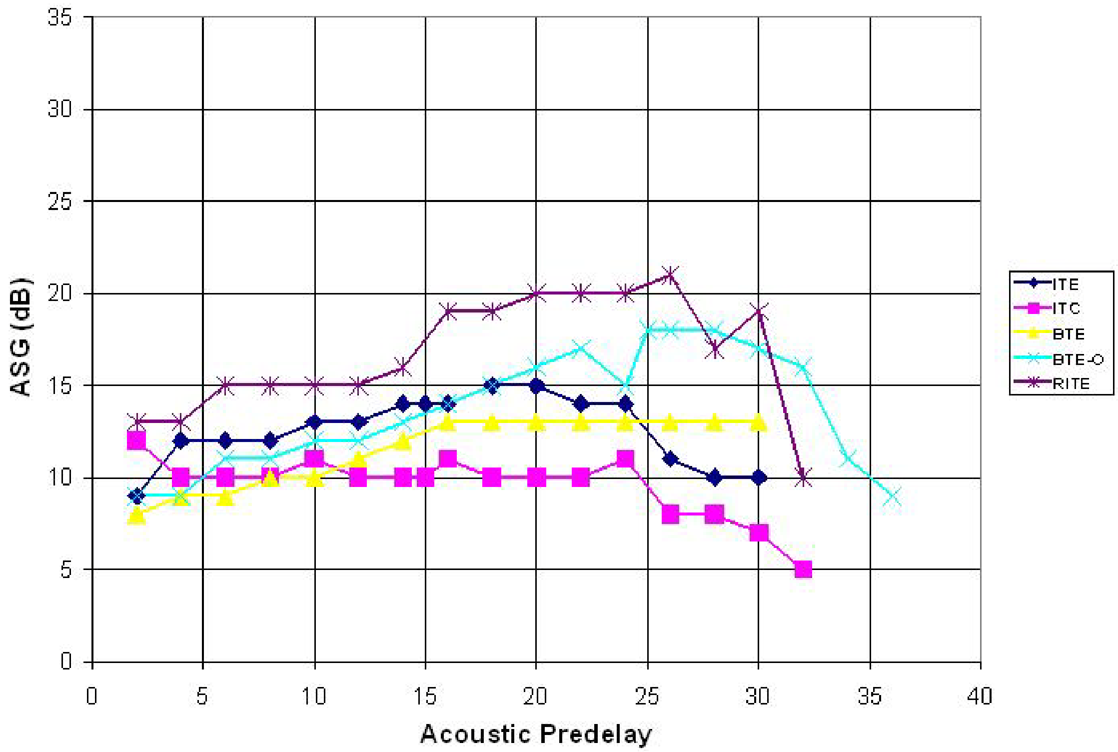

Figure 4. Added stable gains measured for various hearing aid styles as a function of acoustic predelay for a 32 kHz sampling frequency.

This exercise was repeated for each hearing-aid style and for both standard audio sample rates. The maximum ASG measured in each case is provided in Table 1. As shown, AFC3 provides up to 30 dB of ASG, depending on hearing-aid style and audio sample rate.

Table 1. Measured added stable gain for different hearing aid styles and standard sampling frequencies

Hearing Aid Style	Maximum ASG (dB)	
	16 kHz	32 kHz
ITC	11	12
ITE	19	15
BTE	13	13
BTE-Open	18	18
RITE	30	21

These measurements represent a single hearing aid within each style category. Individual results will vary. It is best to perform an acoustic-predelay optimization procedure on any new hearing-aid style to which the ON Semiconductor' feedback canceller is being applied. As a starting point, however, ON Semiconductor recommends the acoustic pre-delay settings shown in Table 2 Note that the acoustic pre-delay is different for the 16 kHz and 32 kHz sampling frequencies. In fact, the acoustic pre-delays for 32 kHz are approximately double of those for 16 kHz.

Table 2. Recommended acoustic pre−delay values for different hearing−aid styles

Hearing Aid Style	Recommend Acoustic Predelay		Receiver to Mic Distance (cm)
	16 kHz	32 kHz	
CIC	7	14	2
ITC	8	15	2.5
ITE	8	15	3.5
RITE	10	20	8
BTE	13	25	14

Acoustic Response

Even for hearing aids of the same style, significant performance differences can be observed. The reason for this is that the feedback canceller effectively removes a limited segment of the impulse (time) response of the feedback path. For feedback paths with a short time duration, the feedback canceller can remove a large proportion of the total energy; for feedback paths with a long time duration, the energy removed is a smaller proportion of the total, resulting in a lower ASG.

To illustrate, the impulse responses of two ITE-style hearing aids are shown in Figure 5. Since the physical sizes of the two devices are similar, they possess similar onset times; however, the response shown in the top graph has a much longer time duration than the response shown in the bottom graph. The feedback-path frequency responses for the same systems are shown in Figure 6. It can be seen in Figure 6 that the first ITE with the longer time duration impulse response is manifested as a very “peaky” frequency response plot.

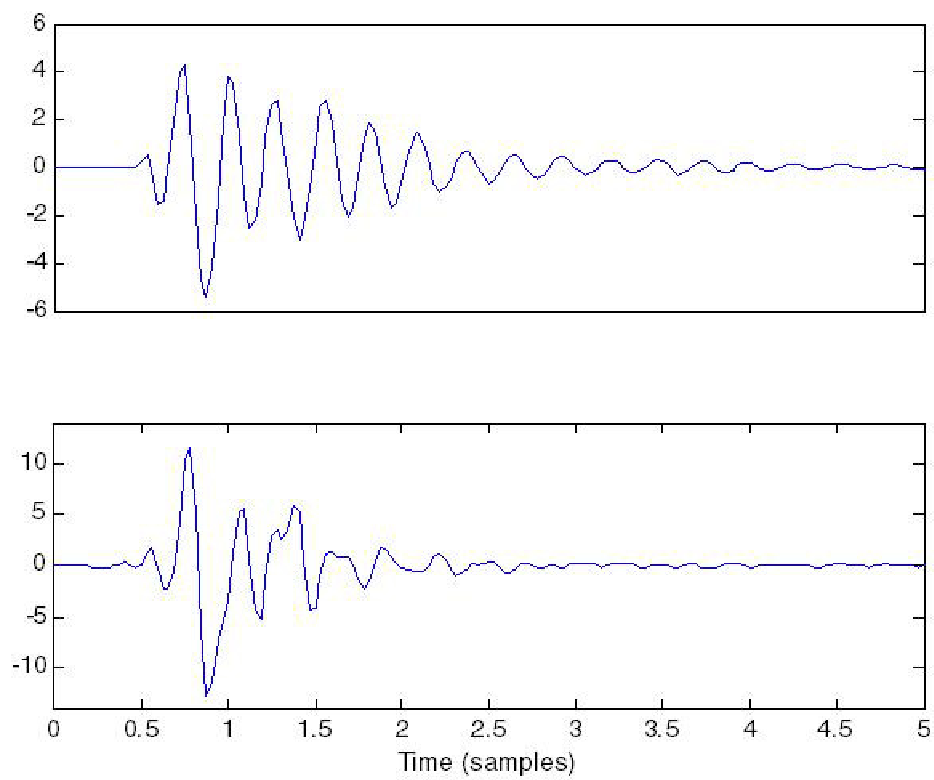

Figure 5. Impulse response for two different ITE hearing aids. Response shown in the top graph has a longer time duration and will likely result in less ASG compared to the response shown in the bottom graph.

There are many factors that can contribute to such peaky feedback path responses: sharp resonances in transducer responses (most notably receivers); acoustic tube resonances and vibration coupling problems. All of these can lead to peaky feedback responses and, ultimately, to poor feedback canceller performance. For optimum feedback canceller performance, such resonances should be eliminated from the system design, as much as possible.

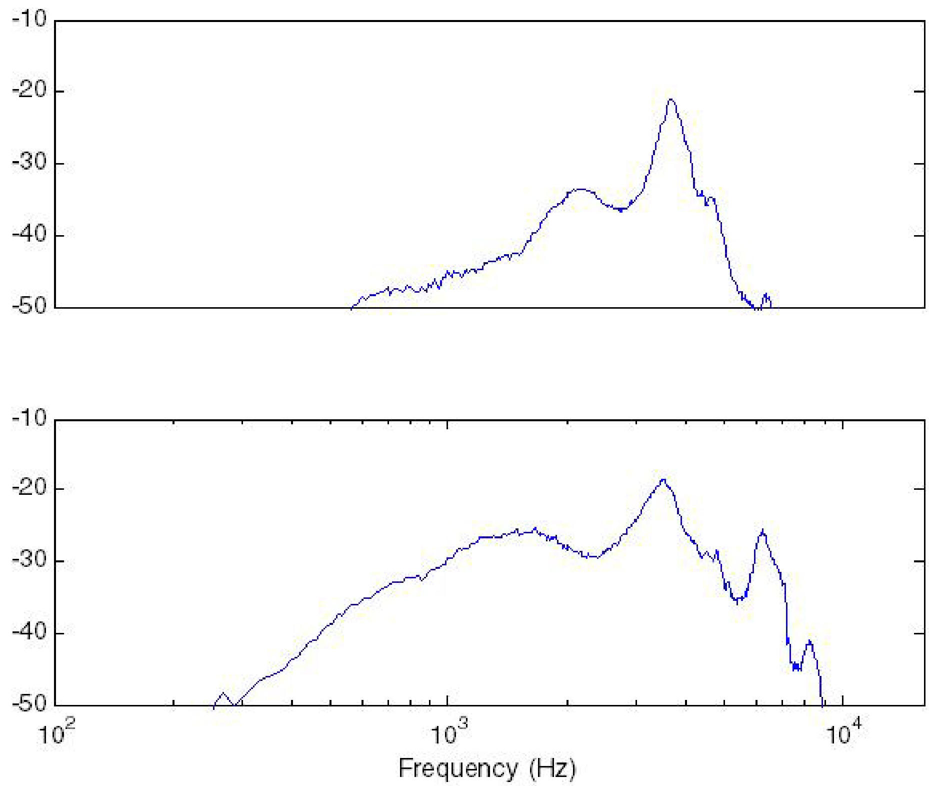

Figure 6. Frequency response for the two ITE hearing aids shown in Figure 5. Response shown in the top graph has a sharper response peak and will likely result in less ASG compared to the response shown in the bottom graph.

Audio Signal Bandwidth

Digital amplifiers from ON Semiconductor offer the flexibility of operating at an audio sample rate of either 16 or 32 kHz, providing audio bandwidths of 8 and 16 kHz, respectively. While the feedback canceller supports both sample rates, some slight differences in performance might be observed.

As described above, the internal feedback path model is only capable of representing a finite time segment of the actual feedback-path response. This finite-time segment consists of a fixed number of audio samples regardless of the sample rate. As the sample rate is lowered, the sample period increases and, for a fixed number of samples, the canceller time window is longer. A longer canceller time window means more of the feedback path can be internally modelled and cancelled, as in Figure 1. This generally leads to better feedback canceller performance (higher ASG).

As an example, the ASG as a function of acoustic predelay was measured for an ITE hearing aid using both the 16 and 32 kHz sample rates. The results are shown in Figure 7. As expected, the lower sample rate provides an increase in the time window of the feedback canceller and results in a higher, peak ASG.

In general, it is reasonable to expect higher ASG for a given hearing aid when the DSP is run at a lower sample rate. Due to acoustical variances between different hearing aid designs (as shown in Figures 5 and 6), however, the same benefits may not be observed in all cases.

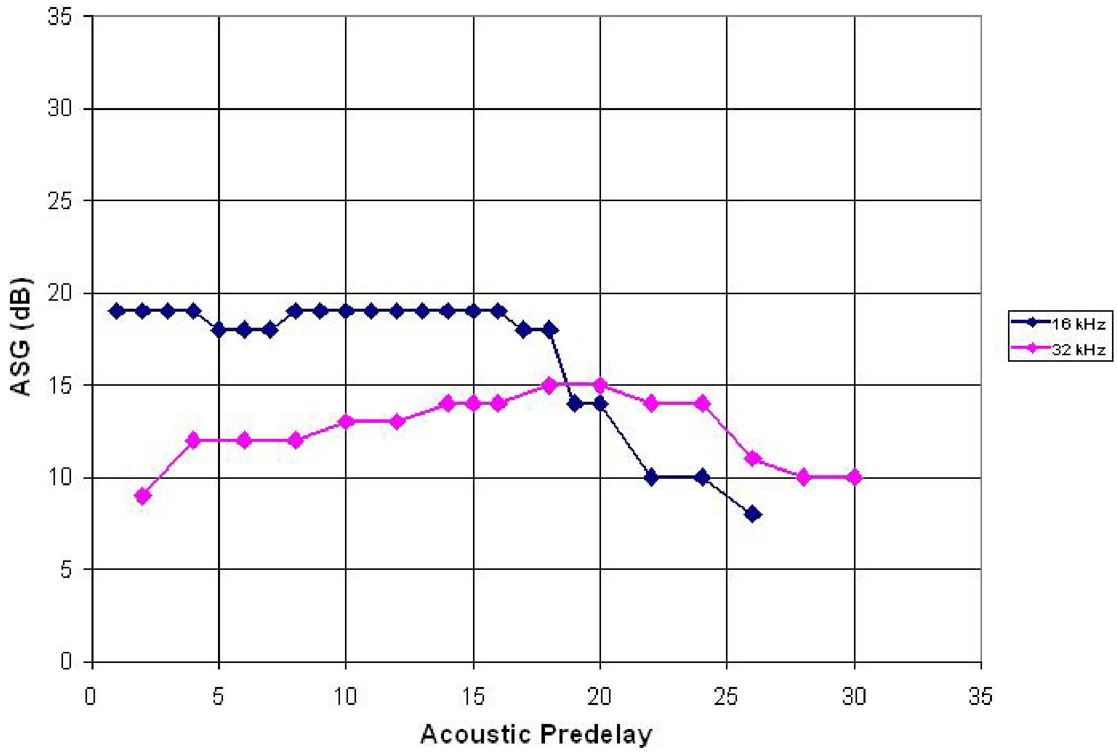

Figure 7. Measured added stable gain for an ITE at two different audio sample rates.

Input Signal Characteristics

In order to minimize disruption for the hearing-aid wearer, feedback canceller updates are performed while the hearing aid is operating. Unfortunately, the adaptation algorithm that is used to update the internal model is sensitive to the microphone-signal characteristics. The adaptation algorithm is ideally trained using non-tonal or noise-like signals. Signals that are tonal, or music-like, can cause maladjustment of the internal feedback-path model leading to renewed feedback or other audible artefacts. This maladjustment phenomenon is sometimes referred to as entrainment.

AFC3 is specifically designed to minimize entrainment artefacts. A sophisticated signal analysis scheme coordinates the adaptation control to permit rapid identification of the feedback path during ideal conditions but to slow adaptation when tonal signals or music are detected. The result is a substantial reduction of entrainment artefacts for music and tones when compared to previous products.

Interaction with Other Hearing Aid Features

An adaptive feedback canceller requires a certain amount of time to adjust itself to any changes in the external environment. While feedback cancellers typically respond to slow variations that occur naturally during hearing aid usage, advanced hearing-aid features may cause much more rapid changes to occur. For example, an adaptive directional microphone can cause a sudden change in the external feedback path since it is rapidly changing the microphone's directional response. Similarly, an adaptive noise reduction algorithm may cause a very rapid gain change if the classification of a frequency band changes from noise to speech.

If a sudden change occurs in the external feedback path there will be a temporary mismatch with the internal feedback-path model, even though the hearing-aid gain did not change. If the gain is sufficiently high, this may lead to a sudden, but temporary feedback event.

Also if the hearing-aid gain changes too rapidly, even though the external feedback path did not change, a small maladjustment of the internal model can be amplified and can lead to a temporary feedback event.

In light of these potential interactions, AFC3 was specifically designed to coexist with other advanced hearing aid features. Improvements in adaptation accuracy, reduced entrainment together with increased coordination ensure minimal disturbances of feedback canceller operation.

The feedback canceller is enabled using the advanced features tab in ON Semiconductor' Interactive data Sheet (IDS) software, as shown in Figure 8.

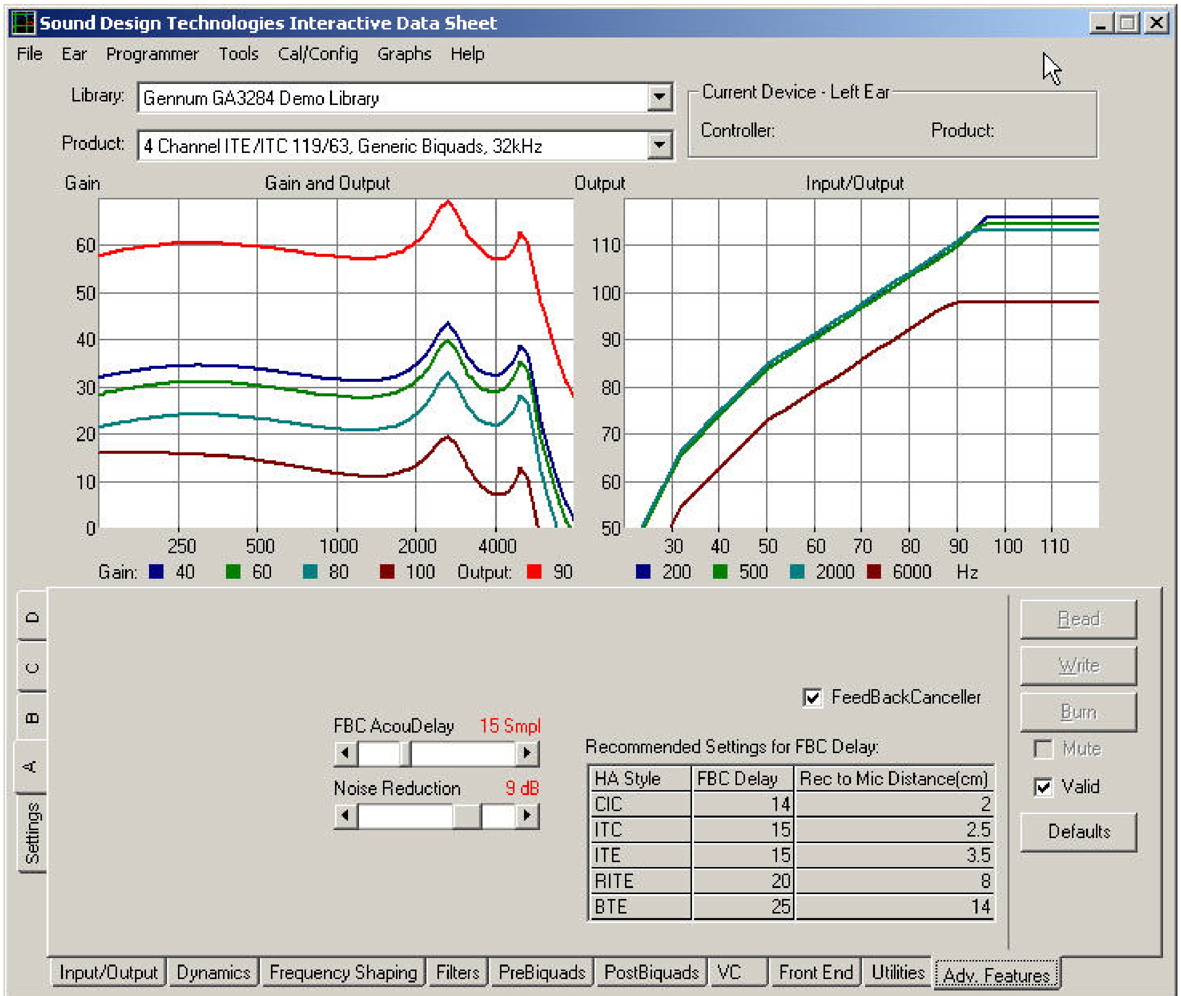

Figure 8. Advanced features tab in ON Semiconductor' Interactive Data Sheet (IDS)

Despite the improvements in AFC3, the unmatched flexibility offered in ON Semiconductor digital hybrids may still result in situations where feedback canceller performance may be impaired. One such case is with the application of the in-channel squelch algorithm.

In certain advanced hybrids, ON Semiconductor offers the ability to apply squelch independently in each compression channel. Experience has shown, however, that having drastically different squelch settings can lead to problems with feedback canceller performance. ON Semiconductor, therefore, recommends that the squelch ratio be restricted to 1:2 and be enabled in all channels when also using adaptive feedback cancellation. Furthermore, the squelch thresholds should be set higher than the microphone noise floor. An example of typical settings is shown in Figure 9.

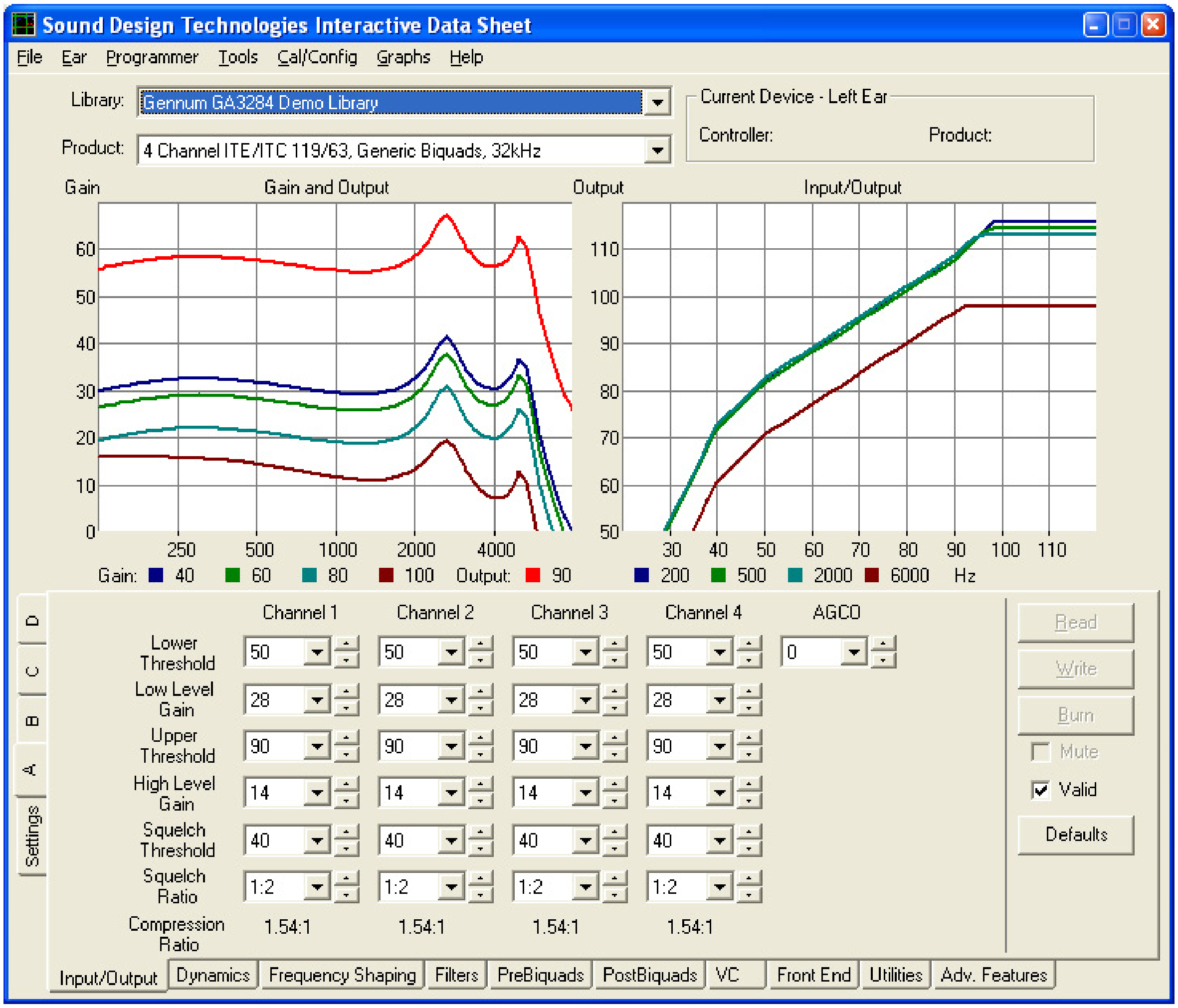

Figure 9. IDS software showing squelch parameters suitable for use with AFC3

SUMMARY

AFC3 from ON Semiconductor is a true feedback cancellation algorithm, removing acoustic feedback without compromising the forward gain of the hearing aid. It achieves added stable gain of up to 30dB while providing immunity to tonal inputs and minimal entrainment artefacts.

The algorithm is designed to coexist with other adaptive hearing-aid features and allows hearing-instrument designers the flexibility to optimize performance for their individual products.

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