Coupling FM Systems to High-Technology Digital Hearing Aids

John A. Nelson, Ph.D.
Widex Office of Research in Clinical Amplification

Hearing aids that use digital signal processing can provide desirable features that could not be realized with conventional ear-level analog hearing aids. While children can benefit from these features, the FM system is still the preferred choice for increasing the signal-to-noise ratio in a classroom. By coupling the two systems together, the benefits of both devices can be obtained. When electroacoustic measurements are made with these combined systems, specific procedures need to be followed to ensure reliable and repeatable results. In this article, hearing aid features that affect electroacoustic measurements will be discussed and procedures will be recommended for obtaining the measurements.

Introduction

The FM system, first introduced in the late 1960s, has become a popular assistive listening device (ALD) in educational settings. The main purpose of an FM system is to transmit an acoustic signal across distance and deliver it to a listener without a reduction in acoustic strength. By accomplishing this, the signal-to-noise ratio (SNR) is typically increased. The positive effect of the increased FM-signal intensity and improved SNR has been presented in many research documents (Boothroyd & Iglehart, 1998; Hawkins, 1984; Picard & Lefrancois, 1986; Pittman, Lewis, Hoover, Stelmachowicz, 1999).

Until recently, the FM system was primarily used as a stand-alone unit. In these FM systems, the FM receiver amplified the transmitted signal and shaped its frequency response. The signal was then delivered to a child's ear using a dedicated receiver, i.e., button receiver, BTE microphone, or headphones. In these cases, the children would remove their personal hearing aids during educational periods. Thus, use of the system was independent of the child's hearing aids.

Most FM systems are single-channel processors with two signal processors: one in the FM transmitter and one in the FM receiver. The FM transmitter is a non-linear system and incorporates a compression threshold to ensure that signal from the microphone will not exceed the dynamic range that can be transmitted by the FM channel. In many systems the compression threshold is measured around 70 dB SPL. Thus, microphone inputs below 70 dB SPL are processed linearly and more intense inputs are compressed with a high compression ratio.

The second signal processor is in the FM receiver and is adjusted for the child's hearing loss. A very flexible system will allow for frequency shaping, gain, and output adjustments. Most FM receivers also incorporate output-limiting compression to decrease distortion of high-output signals. It is important to realize that an FM system typically has two compression systems. The first is in the FM-transmitter and occurs prior to the gain applied for the child. Thus, the FM-transmitter implements input compression meaning the compression threshold is independent of the gain of the FM receiver. This is important to realize, as it will limit the dynamic range of the FM-microphone input, regardless of the gain applied at the FM receiver. The second compression system occurs in the FM receiver and is set to ensure that loud sounds are not uncomfortable or distorted.

On the other hand, developments in hearing aid technology have provided listeners with more advanced signal processing than the traditional single-channel linear and simple compression systems. For example, current compression systems can provide high gain to soft sounds for audibility, less gain for conversational speech, and even less or no gain for loud inputs – all done automatically. Further, when compression is implemented into a multiple-channel system, gain for intense vowel sounds may not affect the gain necessary for the audibility of soft high-frequency consonant sounds.

Many of the obstacles in providing higher technology in a cosmetically acceptable package were overcome with the introduction of digital signal processing (DSP) hearing aids. Digital signal processing may provide features that are not easily achievable with analog hearing aids due to physical size and power consumption limitations. It is important to recognize that not all digital hearing aids provide advanced features. Therefore, one cannot assume a digital hearing aid is superior to an analog hearing aid; it is the processing that improves the hearing aid benefit.

A very low compression threshold (below 40 dB SPL) is one feature that is only available with high-technology digital hearing aids. This feature can provide higher gain for soft sounds resulting in increased audibility. When combining the low compression threshold with a multi-channel hearing aid, the probability of hearing a soft sound, e.g. /f/, is increased. The audibility of these sounds is critical especially when a child is learning speech and language. Other advanced features of digital hearing aids are speech-enhancement algorithms and feedback canceling/management systems. For children and adults, it has been demonstrated that many of these features increase speech-recognition scores and quality of life .

While children often listen and learn in non-ideal signal-to-noise ratios, the benefits of an FM system are still critical. By coupling high-technology hearing aids with FM systems, a child can benefit from both the advanced signal processing and the FM transmission. To further ensure that a signal is delivered without interference, direct audio input (DAI) is the preferred coupling method. When an FM system is coupled to a personal hearing aid, the FM system is responsible for delivering the remote-microphone signal to the hearing aid and is no longer responsible for the processing necessary for the child's hearing loss. When combining two systems, it is important to confirm that the two systems are compatible and provide a high-quality signal to the child.

While some electroacoustic measurements of combined FM and hearing aid systems have been studied , most of the interactions between the advanced features of high-technology hearing aids and FM systems have not been reported. Measurements are necessary to confirm that any interactions are not detrimental to the amplified signal. One way to document compatibility is by validating transparency between the two systems.

Transparency is achieved when inputs to the FM microphone provide the same hearing aid output as those inputs to the hearing aid microphone. Demonstrating this step is critical to deliver a high-quality undistorted signal. System transparency can be documented with frequency response, gain, output, and compression measurements. Advanced signal processing measurements should also be considered when testing for transparency although, more appropriate test signals (e.g., ICRA noise) may not be available on all electroacoustic measurement systems.

This goal of this paper was not to recommend a particular FM+HA fitting method (ASHA, In Press), but to provide insight into the considerations necessary to document the interaction between the hearing aid and FM system. This paper will demonstrate and discuss some of the interactions between these systems and the electroacoustic measurement tools. Recommendations will be made on how to ensure the measurements are stable and reliable. After transparency is documented for a FM+HA system, the desired FM+HA fitting method can be applied.

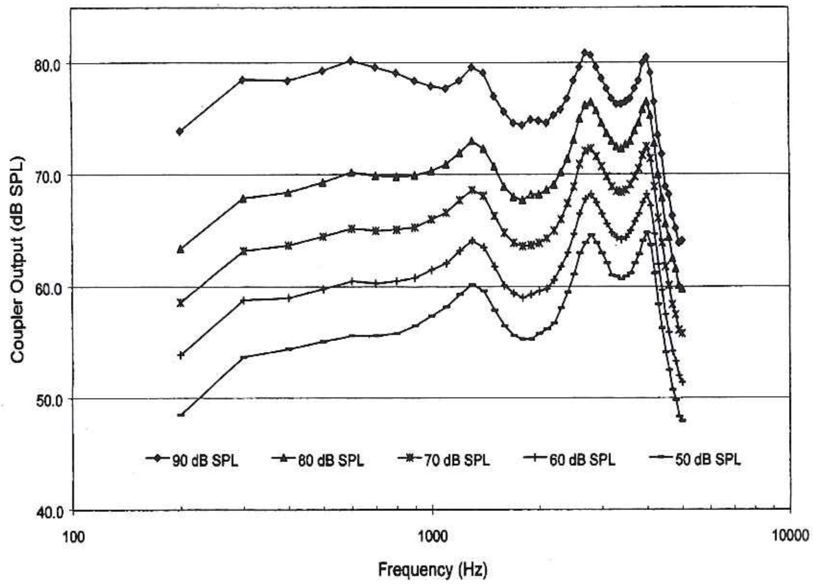
Documenting Transparency

__Frequency Response: Gain and Output Settings__

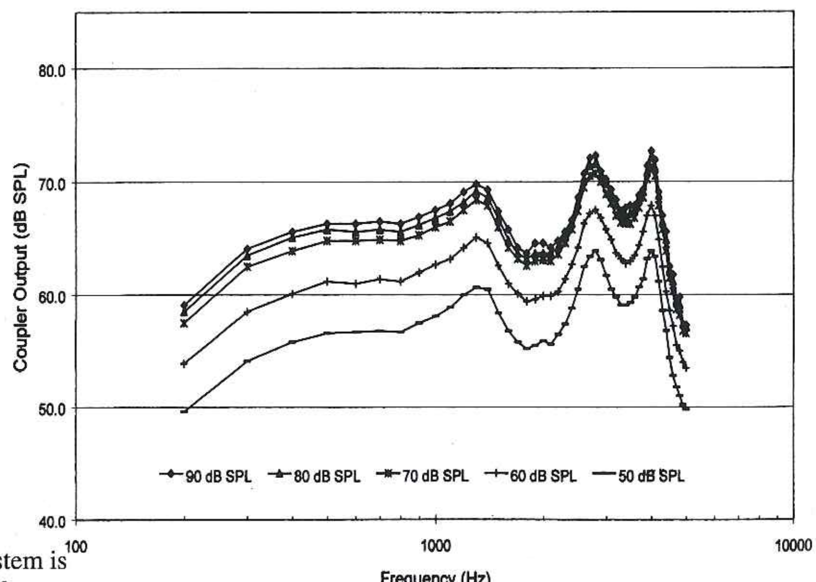
The first step necessary to determine if an FM+HA system is transparent is to evaluate the frequency responses at multiple input levels. Figure 1a is the 2-cc coupler frequency response for the hearing-aid only (HA-only) condition for input intensities of 50 – 90 dB SPL. These measurements indicate that the hearing aid is non-linear (i.e., a 10 dB increase in input corresponds to less than 10 dB change in output). Figure 1b shows the 2-cc coupler frequency response curves for an FM system coupled to the same hearing aid (FM+HA condition). If the FM system is transparent, the FM+HA frequency response curves will be identical to the curves measured in the HA-only condition. For the measurements shown in Figure 1, this is true only for inputs of 50 – 70 dB SPL.

Figure 1. Frequency response curves as measured in a 2-cc coupler obtained at multiple input intensities for the (a) hearing aid microphone and (b) FM microphone coupled to the hearing aid.

1(a)

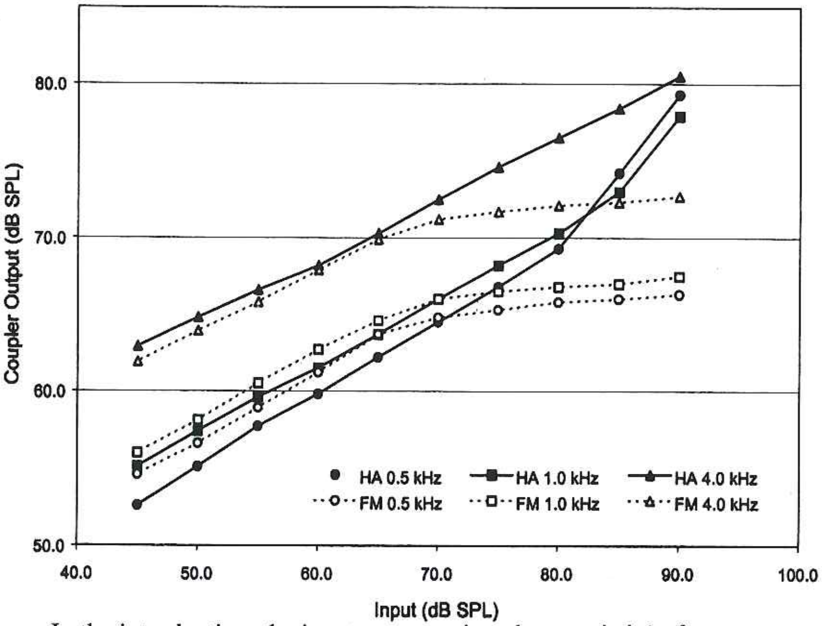


1(b)



To further investigate these differences, input-output curves for 500, 1000, and 4000 Hz were extracted from the previous data (Figure 2). The input-output curve is commonly used to demonstrate the compression characteristics of a hearing aid. Again, the FM+HA deviation from HA-only occurs for inputs above 70 dB SPL. The differences in these outputs can be explained by the input-output characteristics of the FM transmitter.

Figure 2. Input-output curves at 500, 1000, and 4000 Hz for signals processed through the hearing aid and hearing aid coupled to an FM system.



In the introduction, the input compression characteristics of the FM transmitter were discussed. For the systems that the author has tested, this typically results in an FM system that is linear below about 70 dB SPL and has a high compression ratio above this compression threshold. Therefore, transparency between the HA-only and FM+HA is limited to inputs below this compression threshold. The negative side to this FM compression threshold is that intensity changes above the compression threshold are not reflected in the hearing aid output in the same way the hearing aid microphone signal would be amplified. For example, if the teacher's voice rose from 70 to 80 dB SPL at the FM microphone, the hearing aid output would increase minimally.

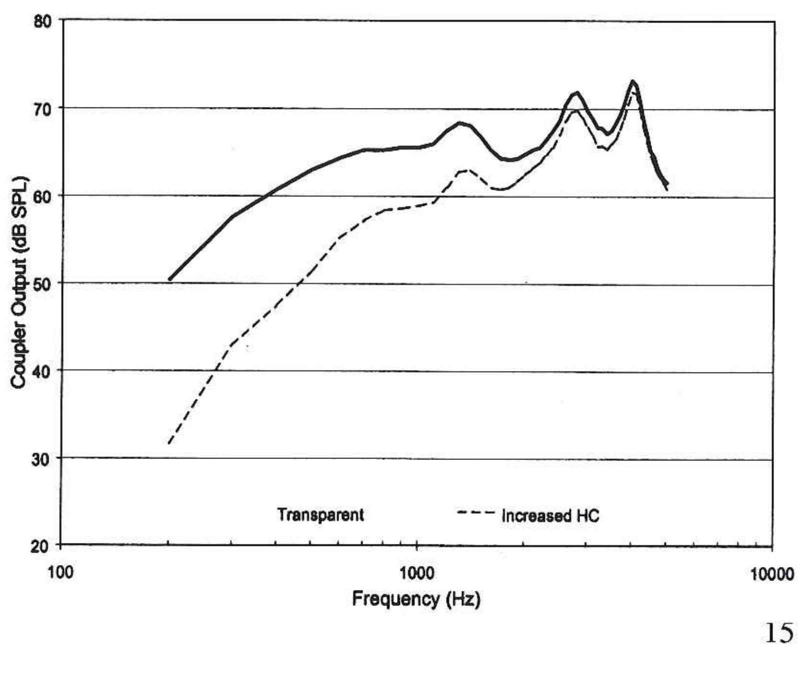
When adjusting a stand-alone FM system, the gain of the FM transmitter would be adjusted to ensure that the FM input was within the child's auditory dynamic range. When an FM system is directly coupled to a child's hearing aid, it is necessary to ensure that the hearing aid will amplify the FM-transmitted signals to be audible and within the child's auditory dynamic range. Further, the audiologist is cautioned that the FM transmitter may utilize compression characteristics other than that of the child's hearing aid. For example, if the FM transmitter uses fast-acting compression, the amplitude fluctuations of speech could be decreased and potentially eliminate the temporal cues helpful in understanding speech. On the other hand, some high-technology digital hearing aids implement slow-acting compression to retain the temporal dynamics of the signal. Some children may prefer the hearing aid microphone with these dynamics to the FM system without these dynamics. Unfortunately, few systems allow for adjustments to the compression characteristics of the FM-transmitter.

The data collected for Figures 1 and 2 were for a non-linear hearing aid and an FM system that does not have adjustable options. The FM system and coupling was developed to be transparent. The benefit of this type of system is that the audiologist does not need to worry about making adjustments to the FM system to meet transparency. When an FM system with adjustable gain, output, and frequency response is used with the direct audio input (DAI), the FM adjustments must be carefully set. This is necessary to ensure that the desired output signal is obtained.

In the next example, an FM system that was not manufactured to be transparent with hearing aids. Using a 65 dB SPL speech-weighted composite noise, the hearing aid frequency response was obtained. This measurement will be the reference for FM-system transparency. The FM system was then coupled to the hearing aid and the same 65 dB SPL speech-weighted composite noise was used as the input for the FM system. The FM-receiver gain was adjusted until the output of the hearing aid was the same as for an input to the hearing aid microphone. Thus, transparency was obtained for the 65 dB SPL stimulus. The transparent frequency response curve is shown in Figure 3.

Following this transparency setting, multiple-level frequency response curves should be obtained as discussed previously. Adjusting the FM gain at various intensity inputs should be done with caution. These gain adjustments will typically affect the response curves for other intensity levels. Two cases will now be described where inaccurate FM adjustment can negatively alter the FM+HA transparency and also the amplified signal delivered to the child's ear.

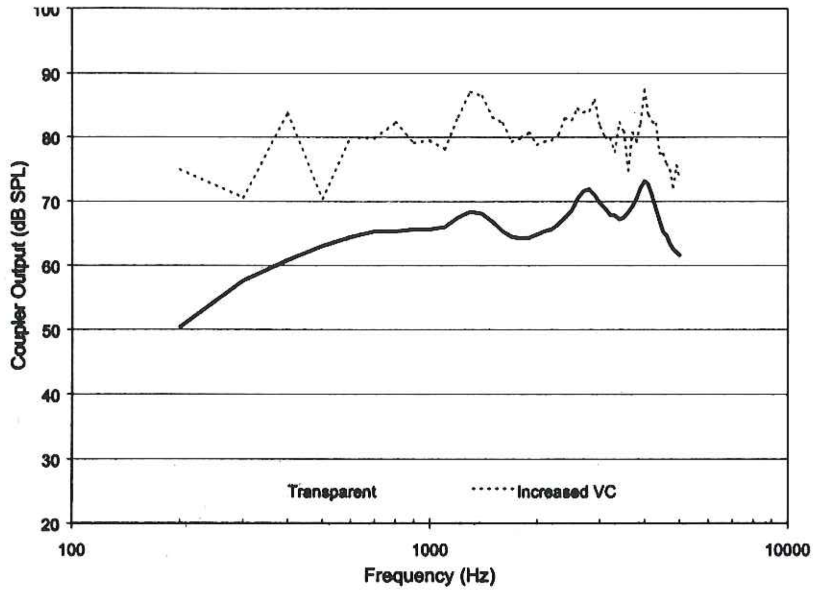
Figure 3. Frequency response curves measured in a 2-cc coupler obtained for an FM system coupled to a hearing aid. The curves are for an FM+HA system that is transparent and with the FM low-cut tone control activated.



When adjusting a stand-alone FM system for children with high-frequency hearing losses, the FM receiver would be adjusted for high-frequency gain emphasis. When coupling an FM system to a hearing aid, this same adjustment may not be appropriate. It is important to remember that the hearing aid has been programmed for the child's hearing loss and will amplify and shape the frequency response based on that program. Thus, the frequency shaping of the output signal is processed within the hearing aid. In order to obtain transparency when using an FM+HA system, the FM receiver should be set for the broadest frequency response. The dashed line of Figure 3 demonstrates the non-transparent frequency response for an increased low-cut (LC) of the FM system. In this case, the low frequencies are filtered twice: first by the FM receiver and second by the hearing aid.

Electroacoustic measurements are also necessary when adjusting the gain/output of the FM+HA system. Some FM+HA fitting methods suggest that the FM system gain should be increased for an improved signal-to-noise ratio for the FM signal (ASHA, In Press). For some FM+HA combinations, increasing gain to meet these FM+HA fitting methods can be detrimental. The dotted line in Figure 4 demonstrates the FM+HA output when the FM volume control of the previously transparent system was increased. Notice that the output is distorted compared to the transparent settings. Specifically, the curve is not smooth and does not follow the shape of the output generated by the hearing aid microphone. If FM+HA system were capable of transmitting the signal with increased FM gain, the output would be the same shape as the hearing aid microphone curve but shifted vertically on the graph. This measurement indicates that the increased output of the FM system is distorting the input to the hearing aid DAI. It is again strongly recommended to obtain electroacoustic measurements after all adjustments to either the hearing aid or FM system.

Figure 4. Frequency response curves measured in a 2-cc coupler obtained for an FM system coupled to a hearing aid. The curves are for an FM+HA system that is transparent and with an increased FM volume control setting.



Hearing Aid Compression Characteristics – Attack and Release Times

In the previous section of this article, the effects of the FM system compression threshold and ratio on transparency measurements were discussed. The hearing aid compression threshold and ratio was demonstrated as a non-linear growth of frequency response curves (see Figures 1 and 2). The compression time constants (attack and release times) will also affect the expected coupler measurements. If a hearing aid has very fast time constants, the gain quickly adjusts to changes in input intensity. In this situation, the coupler measurements will probably be consistent for repeated measurements.

When slow-acting compression is implemented, the gain will change slowly for changes in input intensity. In these situations, the electroacoustic measurement will be dependent on the duration of the stimulus presentation prior to recording the output. This is a key point in documenting transparency as comparisons are made across multiple measurements and the hearing aid needs to be providing the same signal processing for each comparison.

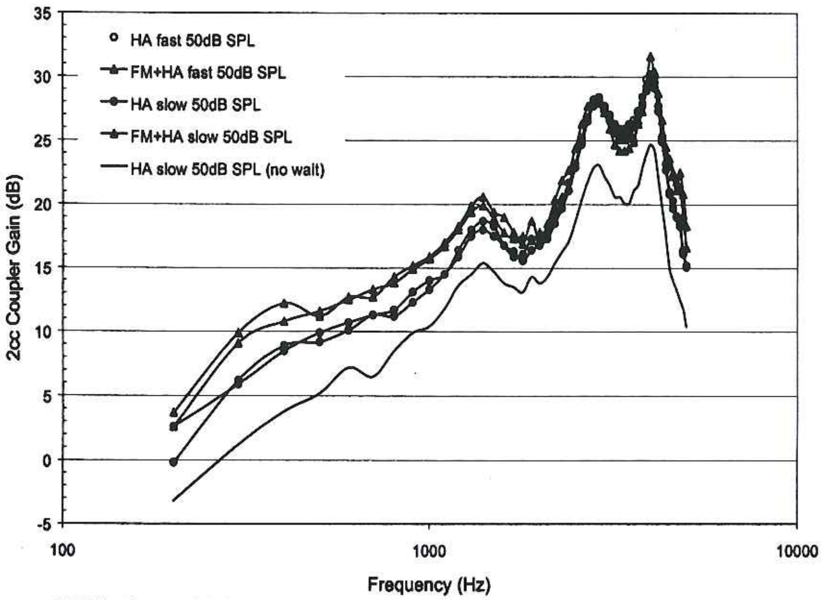
Consider an example where a hearing aid is providing gain of 5 dB at 2000 Hz for a 65 dB SPL signal. The input signal is quickly reduced from 65 to 50 dB HL and the hearing aid realizes that gain should be increased to 12.5 dB at 2000 Hz. This corresponds to a 2:1 compression ratio. To preserve the temporal cues, a slow-acting compression system might take 20 seconds to stabilize at the new gain setting. This is also similar to an individual who adjusts a volume control based on average changes in the environment. If a coupler measurement is recorded during that 20 seconds, the gain values will vary between 5 – 12.5 dB. Fortunately, if the measurement is collected after these 20 seconds of continuous presentation at 65 dB SPL, the hearing aid gain will be stable.

The response curves can also be used to demonstrate these effects and are in Figure 5. The hearing aid was programmed for a mild-to-moderate sloping hearing loss (HA condition) and the FM system was pre-set for transparency (FM+HA). Each curve was obtained with a 50 dB SPL composite noise signal in a hearing aid test box. Four of the curves overlap on the top of the graph (higher gain). Two of these curves are for the HA condition: one obtained with the hearing aid set with fast time constants and one obtained with the hearing aid set with slow time constants. For the slow-acting compression, the measurement was obtained after the composite noise was active for more than 20 seconds and the hearing aid gain had stabilized. The other two overlapping curves were for the same conditions, except the input signal was to the FM microphone (FM+HA condition).

The last curve on Figure 5 demonstrates the potential measurement variability using slow-acting compression (see HA slow 50dB SPL, no wait). In this case, the hearing aid microphone was provided a more intense signal for 20 seconds, long enough for the hearing aid gain to stabilize for that input level.

The composite noise was then quickly decreased to 50 dB SPL and the frequency response was obtained before the slow-acting compression stabilized (gain was increasing). This measurement is a reflection of the slow-acting compression system and demonstrates a measurement artifact. Therefore, the timing of the HA and FM+HA measurements is critical to ensure that the hearing aid is operating with the same gain characteristics for both HA and FM+HA measurements.

Figure 5. Frequency response curves as measured in a 2-cc coupler obtained for a hearing aid and hearing aid + FM system with slow and fast hearing aid compression time constants.



While it would be ideal to temporarily adjust the time constants for fast-acting compression during coupler measurements, the option may be unavailable in the programming software or the software might not be available to the individual making the measurements. In these cases, it is important to measure the frequency response curve after the gain has stabilized. This duration varies across hearing aids and is dependent on the longest attack and release time. For many high-technology hearing aids, the gain stabilizes after presenting a stimulus with steady amplitude for 20-30 seconds.

Speech Enhancement / Noise Reduction Algorithms

Speech enhancement / noise reduction algorithms analyze the temporal characteristics of the input signal and reduce the gain if the characteristics are similar to those of noise-like signals. Typically, the analysis is done in individual hearing aid channels or frequency regions. For example, low-frequency speech sounds, e.g., vowels, fluctuate in amplitude over time at the rate of the speaker's fundamental frequency. A continuous low-frequency noise created by an air conditioner would have minimal amplitude fluctuations over time. The speech enhancement algorithm determines the input (noise) does not have speech-like fluctuations and decreases the gain in that frequency channel.

In many typical listening situations, speech enhancement algorithms can be viewed as a benefit. However, the test stimulus for electroacoustic measurements is often a pure-tone sweep or composite noise with steady-state amplitude. When using this stimulus for testing hearing aids with speech enhancement algorithms, special considerations need to be made for the data collection and analysis.

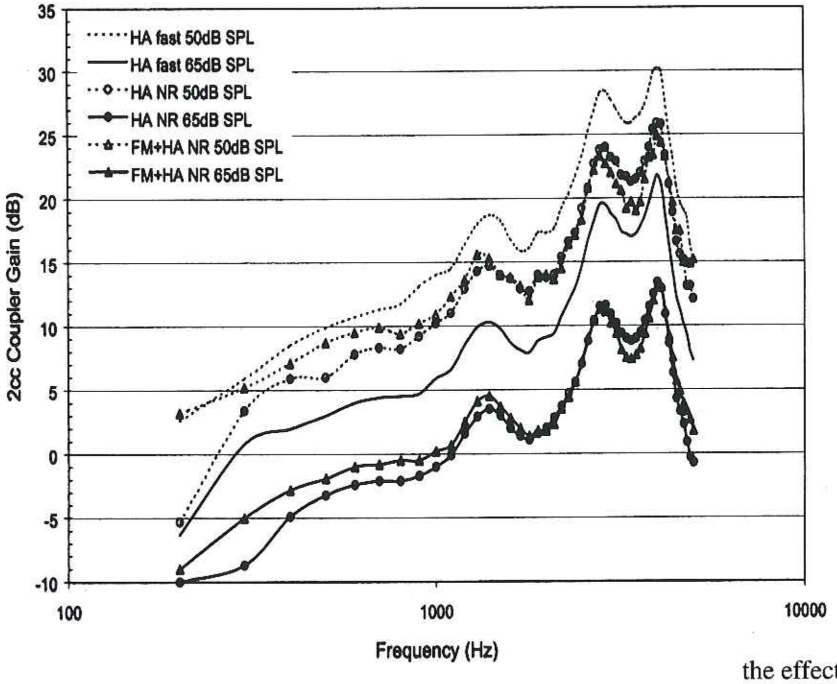
To decrease artifacts due to inaccurate identification of the input signal, speech enhancement algorithms analyze the input signal over a period of time prior to adjusting the hearing aidgain. Thus, it is a slow-acting system. Based on the discussion of slow-acting compression systems in thia article, it is reasonable to assume that if a continuous composite noise is presented for a long duration the gain would stabilize and consistent measurements could be obtained. Based on the current algorithms, this assumption holds true.

To demonstrate results of one speech enhancement / noise reduction system, multiple frequency response curves were obtained at two input levels: 50 and 65 dB SPL. Figure 6 presents these response curves. After temporarily deactivating the noise reduction, hearing aid response curves were obtained for the two input levels (HA-fast 50 dB SPL; HA-fast 65 dB SPL). Then, the noise reduction algorithm was activated and response curves were obtained for HA and FM+HA configurations. To ensure stable gain from the hearing aid, these measurements were obtained after the composite noise was active for more than 30 seconds and the hearing aid automatically adjusted the gain based on the input signal. The hearing aid manufacturer recommended this duration based on the speed of the speech enhancement algorithm. If measurements were obtained prior to stable gain, i.e. noise reduction in full effect, the response curve will vary somewhere between the stable value and the fast-acting curve. From these curves it was concluded that transparency could be documented if the hearing aid speech enhancement algorithm was fully activated and gain was stable.

Further investigation of Figure 6 indicates that the noise reduction algorithm decreased the gain by approximately 2.5 dB for the 50 dB SPL input signal and 7.5 dB for the 65 dB SPL input signal. Recall that these response curves for noise-like inputs and not speech-like inputs. If a speech-like stimulus was used, the fast-acting hearing aid gain values would be recorded. Further, the speech enhancement algorithm in this hearing aid is non-linear and the degree of gain reduction is dependent on the input intensity. As a result, these measurements can be used to determine FM transparency but cannot be used to verify gain for speech stimuli.

If the noise reduction systems can be temporarily deactivated, transparency measurements can be obtained without special precautions. Otherwise, measurements should be made after the hearing aid gain is stable, approximately 30 seconds of signal presentation. Another alternative is to use special digital speech stimuli that have been created with amplitude fluctuations that are similar to speech. (For a more detailed description of

Figure 6. Frequency response curves as measured in a 2-cc coupler obtained for a hearing aid and hearing aid + FM system with a hearing aid speech enhancement algorithm activated (NR) and deactivated (fast).



these stimuli please see Dreschler, Verschuure, Ludvigsen, & Westermann, 2001.) The hearing aid algorithm will determine the input to have speech-like temporal characteristics and the gain will reflect the gain applied for speech inputs.

Internal Noise of the Combined Hearing Aid and FM System

The internal noise of a hearing aid is typically measured as the equivalent input noise level (ANSI S3.22, 1996). This measurement is the average output at 1000, 1600, and 2500 Hz for a 60 dB SPL input signal (Lav) and the output for no acoustic input (L2). The equivalent input noise level (EINL) is calculated as

$$EINL = L2 - (Lav - 60) \text{ dB.}$$

In the case of the FM+HA system, the combined internal noise of the system is greater as circuit noise from two systems is processed being processed by the hearing aid. Unfortunately, for compression hearing aids, the EINL measurements are not accurate representations of the noise floor. Specifically, hearing aid gain for the 60 dB SPL input (Lav measurement) will be less than the hearing aid gain for the quiet measurement (L2 measurement). Thus, the noise for the L2 measurement may be higher due to the increased gain for soft inputs. For both measurements, the FM-system gain is usually constant – measurements are obtained below the 70 dB SPL compression threshold. The interactions of these effects increase the difficulty to evaluate the EINL for combined FM+HA systems.

For internal noise measurements, it is more important to recognize the source of the noise and how it is amplified. High-technology digital hearing aids incorporate special processing to reduce the amplification of internal circuit noise while still providing high gain for soft input signals e.g., expansion below the low compression threshold (Kuk, 1999). Currently FM systems are analog signal processors and the internal noise of the system is directed to the hearing aid DAI. If the FM-system circuit noise is above the expansion region of the hearing aid, the noise will be amplified with high gain. In this case, it is hoped that the amplified FM system internal noise will be below the child's threshold of audibility. Unfortunately, this is not always the case.

To demonstrate the effects of internal noise for the two listening conditions, the output of a hearing aid was measured in a coupler placed in a double-walled sound-treated room. No acoustic stimulus was presented in the measurement room – a very quiet environment. A high-technology digital power hearing aid was set for high-gain, necessary for an individual with a profound loss to hear soft-speech in a quiet environment. The hearing aid output was recorded at 90 dB SPL (amplified ambient noise and hearing aid circuit noise). The power hearing aid was then coupled to an FM system that had been set for transparency (i.e., no increased signal amplitude for the FM microphone). The output was recorded at 105 dB SPL (amplified ambient noise and combined FM+HA circuit noises).

A listening check of this FM+HA system by an audiologist with normal hearing will be able to confirm the effect. First, listen to the noise levels from the HA-only with the volume turned high, i.e., in a very quiet environment. Second, listen to the noise levels for the FM+HA system at the same volume setting. It is important to realize that this noise may be below audibility for the child. Then, listen to the noise levels when the microphone input is increased (i.e., turn on another stimulus like a radio or someone talking). For compression hearing aids, this increased input at the microphone will decrease the hearing aid gain. The gain reduction will decrease the output level of the FM-system noise. In both cases, the signal-to-circuit-noise ratio is increased when the hearing aid gain is decreased.

The internal noise of an FM system cannot be lowered without the development of higher-quality systems. Until then, a few suggestions can be made to alleviate these noise problems. One recommendation is to reduce the hearing aid gain for soft speech such that the FM+HA system noise is not audible. For compression hearing aids, this might involve reprogramming the hearing aid. This solution does compromise the audibility of soft inputs, including speech. Another recommendation is to not use the FM+HA system when the speaker is near the listener when listening in a quiet environment. Recall the function of the FM system is to transmit signals over distance without a reduction in signal strength and thus, increasing the signal-to-noise ratio. In quiet listening environments, the SNR is high and the FM-benefit might not be necessary. Finally, it is important to teach older children the different listening options so that they can make informed choices for their amplification needs.

Summary

In this paper, a number of procedural issues of testing FM transparency were discussed. While each electroacoustic measurement was valid, the interpretation of these measurements must be made with caution. The data in this article were obtained with steady-state speech-weighted composite noise. While pure-tone sweeps are commonly used for electroacoustic measurements, previous investigations have recommended that they not be used to evaluate compression hearing aids. Specifically, non-linear hearing aids were developed to process speech signals and measurements obtained with pure-tone signals will not accurately reflect the hearing aid processing (Dreschler et al., 2001). Thus, the author does not recommend using pure-tone signals to evaluate FM+HA systems.

Digital speech signals, e.g. ICRA noise, can also be used to avoid the measurement artifacts of the speech enhancement / noise reduction algorithms. The ICRA signal is a noise that was developed to have both the average spectral and temporal characteristics of speech. However, slow-acting compression will still require digital speech signals to be presented continuously until the hearing aid has reached the prescribed gain for the input level. In summary, the following steps are recommended when making electroacoustic measurements with an FM system coupled to a hearing aid.

Temporarily deactivate high technology special processing like speech enhancement / noise reduction algorithms and adjust compression characteristics to fast time constants. This will reduce artifacts of stimulus type and presentation duration. This would require the hearing aid programming equipment and training.

If the hearing aid cannot be temporarily reprogrammed, obtain measurements with composite or digital speech noise presented for a long duration, until the hearing aid reaches stable gain. Typically this is around 20-30 seconds depending on the hearing brand and model.

Obtain measurements at multiple input intensities for both the HA only and FM+HA configurations. Make adjustments as necessary. Remember that an adjustment may affect the results of a prior measurement. Therefore, confirm all measurements after the final settings are selected.

Use caution when interpreting electroacoustic measurements for desired gain/output for the individual's hearing loss. This is for two reasons (a) special processing might be changing gain based on the characteristics of the input signal (speech or noise like) and (b) generic fitting algorithms do not always apply directly to specific signal processing hearing aids.

Hearing aid technology has developed in two directions: increased cosmetic appeal and increased signal processing. FM systems have increased in cosmetic appeal over the past ten years. They are smaller, lighter, and more stylish in color and shape than previous models. Hopefully, the next generation of FM systems will follow the second direction that hearing aids have taken i.e., increased signal processing. High-technology FM systems could provide higher quality signals, increased dynamic range, and decreased internal noise. Until this time arrives, it is important to realize the limitations of FM+HA interactions and to recommend FM systems when the potential for acoustic signal degradation over distance is a concern or to increase the SNR of the listening environment.

Note: The data presented in this study were obtained using a Frye Electronics Fonix 6500 Hearing Aid analyzer and WinCHAP. The hearing aids were Widex Senso C8+ or P38 coupled to either a MicroLink by Widex or a Phonic Ear PE 475.

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