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1.0 Introduction

This chapter describes electro-acoustical coupler and real-ear measurements of hearing aids using a hearing aid analyzer. There are several reasons to perform these measurements including verifying that the hearing aid amplification is appropriate for the patient’s hearing loss, determining the characteristics of the hearing aid, and checking that the hearing aid is working properly.

There are two different classes of hearing aid measurements: real-ear and coupler. Real-ear measurements (REMs) are performed with the microphone placed inside the ear canal of the patient. REMs are also sometimes known as “probe” measurements, referring to the small probe attached to the microphone that is placed in the ear. The ear canal has a significant impact on the frequency response of the amplification produced by the hearing aid, and every ear canal is unique. Therefore, using REMs is the only way for clinician to determine exactly how much amplification the patient is receiving.

Coupler measurements are performed inside a sound chamber using a device called a coupler. The coupler works as a substitute for the ear canal, providing a standard-sized cavity into which the amplification produced by the hearing aid is directed, and a mechanical way to attach a microphone to the hearing aid. The primary use of coupler measurements is to specify the characteristics of a hearing aid and to determine if it is functioning properly. Coupler measurements are easier and faster to perform than REMs, but they are not a precise measurement of the frequency response of the hearing aid inside the ear. Although both REMs and coupler measurements are technically “electro-acoustical” measurements, usually that term is reserved for coupler measurements. Coupler measurements are also sometimes called “sound chamber” measurements.

Almost all hearing aids being fit today are digital or digitally programmable with computer-based fitting programs, allowing the clinician to perform extensive adjustments to the hearing aid and display the predicted results of these adjustments on the computer screen. Many clinicians have been lulled by the ease of these programs into assuming that the hearing aid will always perform as the program predicts. However, these programs cannot determine the affect of an irregular ear canal resonance on the hearing aid amplification or diagnose problems resulting from a poorly fit shell or earmold. Nor can they predict errors resulting from problems with the hearing aid programming or the changes that may occur over time to the hearing aid response. Therefore, it is critical for the clinician to perform independent verification of predicted results in order to determine the hearing aid’s amplification.

Coupler measurements can be used in to check that the hearing aid is performing to the manufacturing specifications, to pre-fit a hearing aid before the patient arrives at the clinic, and to establish a baseline that can be used later if the hearing aid comes back into the clinic for repair or adjustment. Coupler measurements are the absolute minimum amount of verification that every clinician should perform with a hearing aid fitting.
REMs are an important next step in verifying the hearing aid fitting. Depending upon the clinical setup, REMs can usually be performed while the clinician adjusts the hearing aid during the fitting process. This has the benefit of immediately showing the clinician how adjustments made in the hearing aid program actually affect the response of the hearing aid inside the ear. These REM records can save valuable troubleshooting time later should any problems with the hearing aid occur.
2.0 Coupler Measurements

A coupler measurement is the electro-acoustical analysis of a hearing aid connected to a coupler inside a sound occluded box known as a sound chamber. Coupler measurements are perhaps the easiest and the most convenient way to perform an independent verification of the response of a hearing aid. Tests are performed in the controlled environment of the sound chamber and do not require the presence of the hearing aid wearer. They require little technical skill on the part of the operator and produce consistent, repeatable results in a short amount of time.

The disadvantage of coupler measurements is that the frequency response achieved with a standard 2-cc coupler can be quite different than the response of the hearing aid inside the ear. Coupler measurements do not typically take into account venting, the shape of the hearing aid, or any irregularities that may exist inside the patient’s ear canal; all of these parameters can have a large impact on the real-ear response of the hearing aid. Coupler tests performed with a 2-cc coupler are especially unrealistic for the measurement of CIC hearing aids. The more traditional BTE and ITE hearing aids that 2-cc couplers were designed to work with are not inserted very deeply into the ear canal. The deeper insertion depths of canal hearing aids create a greater overall amplification (given the same amount of power) than the more traditional styles.

2.1 Test equipment

At the most basic level, a hearing aid analyzer requires a speaker to drive a test signal, a microphone to measure the response, a method to connect the hearing aid to the measurement microphone, and a way to analyze and display the data.

In coupler measurements, the speaker used to drive the test signal is contained in a sound occluded box called a sound chamber. Most sound chamber measurements are performed with the lid of the sound chamber closed, creating a stable sound-occluding environment. As shown in Figure 1, the hearing aid is attached to a coupler that functions as an artificial ear cavity that mildly amplifies the hearing aid response similar to the natural amplification provided by the human ear. The measurement microphone is inserted into the

Figure 1: The measurement microphone is inserted into the 2-cc coupler, which is connected to the hearing aid. A battery pill is used to measure the battery current of the hearing aid.
coupler. The hearing aid/coupler/microphone assembly is placed inside the sound chamber where the measurements are made. A second microphone, called a control microphone, is sometimes used by the hearing aid analyzer to monitor the sound pressure level (SPL) of the signal near the hearing aid’s receiver.

### 2.2 Couplers

There are several basic couplers commonly used for performing sound chamber measurements. According to the ANSI S3.22 standard, Behind-the-Ear (BTE) hearing aids are tested using an HA-2 coupler. In-the-Ear (ITE), In-the-Canal (ITC), and Completely-in-Canal (CIC) hearing aids are tested using an HA-1 coupler. The new “open ear” hearing aid style, commonly called Over-the-Ear (OTE) are sometimes tested with a HA-1 coupler combined with a special coupler device, but other times are tested with an HA-2 coupler. Contact the hearing aid manufacturer for details on how to connect the OTE hearing aid to a coupler in order to replicate ANSI S3.22 test results.

Sometimes other couplers are used in order to obtain a more real-ear like frequency response than is possible with an HA-1 or HA-2 coupler. These couplers include the Zwislocki ear simulator, CIC coupler, and Open-fit coupler.

#### 2.2.1 HA-1 coupler

The ANSI S3.22 standard specifies the use of the HA-1 direct access 2-cc coupler (Figure 2) when testing ITE, ITC, CIC hearing aids, and other hearing aids with attached molds. The sound bore of the hearing aid is sealed directly to the 2-cc cavity of the coupler with putty.

![Figure 2: The HA-1 2-cc direct access coupler is used for testing canal-type hearing aids (ITE, ITC, CIC).](image)
To attach the HA-1 coupler to the hearing aid:

1. Roll some putty into a rod long enough to go around the transmitting end of the hearing aid. See Figure 3a.

2. Bend the putty rod around the canal of the hearing aid, making the resulting “donut” flush with its end. Some users choose to seal the vent of the instrument at this end with a small dab of putty. See Figure 3b.

3. Align the sound opening of the hearing aid to the small hole at the end of the coupler. Look through the open end of the coupler to be sure the sound opening of the instrument is clear of obstructions and correctly placed. See Figure 3c.

4. Seal any vent on the hearing aid with a small dab of putty. See Figure 3d.

5. Complete the acoustical sealing of the hearing aid to the coupler by using a pencil or finger. See Figure 3e.

2.2.2 HA-2 coupler

The ANSI S3.22 standard specifies the use of the HA-2 2-cc coupler (Figure 4) when testing BTE aids. The “hat” attached to the main body of the coupler simulates the earmold and provides a tube used to attach the BTE hearing aid. The total length of the HA-2 coupler, including the #13 tubing, should be 25 mm. The length and the width of the tubing can have a significant impact on the frequency response measurement of the hearing aid.

2.2.3 Ear simulator

The ANSI S3.25 standard specifies an ear simulator that better simulates the characteristics of the ear than the HA-1 and HA-2 couplers. In the ear simulator, higher frequencies have a higher gain than lower frequencies, simulating the impedance of
Hearing Aid Measurements

2.2.4 CIC and Open-fit Couplers

When coupled to an HA-1 or HA-2 coupler, some hearing aid styles produce frequency response measurements that are very different than response the hearing aid will produce inside the patient's ear canal. In particular, CIC and Open-fit hearing aids are not well suited to the 2-cc couplers. The CIC and Open-fit couplers were designed by Frye Electronics to produce more realistic frequency responses than is possible with a standard 2-cc coupler.
When a CIC aid is inserted into the ear, it takes up more space inside the ear than a conventional earmold or ITE hearing aid shell. Just as someone’s voice in a small room sounds louder than the same voice in a large room, a CIC aid will produce more amplification inside an ear than an ITE with the same amount of power. This means that an HA-1 coupler measurement of a CIC hearing aid is not a very realistic representation of the CIC hearing aid’s real-ear response.

The 0.4-cc volume of the CIC coupler (Figure 6) is more like the residual volume found in an ear canal containing a CIC hearing aid. The CIC coupler, combined with some software correction factors, is used to produce CIC coupler measurements that are predictor of its real-ear response. (Its accuracy was verified by Dr. Stephen Martinez in his Ph.D. dissertation in 2000.) Typically, the CIC coupler produces 10-15 dB more gain above 1500 Hz than a measurement with the same hearing aid using an HA-1 coupler. See Figure 7 for a comparison of a CIC hearing aid tested in an HA-1 and a CIC coupler.

**Note:** The ANSI S3.22 standard specifies the use of the HA-1 coupler for testing CIC hearing aids. The CIC coupler cannot be used for checking the hearing aid’s performance against manufacturer specifications.
2.3 Calibration and Setup

1. Prepare the testing environment. This might involve “calibrating” the measurement and the control microphones or “leveling” the sound chamber at the reference point. See the hearing aid analyzer operator’s manual for details.

2. Attach the hearing aid to the correct coupler as described in the previous section.

3. Place the hearing aid in the sound chamber (Figure 1). There is often a recommended “reference point” where the hearing aid microphone should be placed for most accurate and repeatable testing results.

4. Close the sound chamber.

5. Test as desired.

2.4 ANSI S3.22

The most common coupler test is a sequence of measurements to the ANSI S3.22 hearing aid specification standard. These tests produce results that can be compared to a test strip containing the hearing aid’s specifications. Every newly manufactured hearing aid comes with one of these test strips. In many clinics, every hearing aid will be checked against its specifications by performing an ANSI test sequence as the aid comes in the door.

The ANSI S3.22 standard has been revised several times throughout the years; the two most recent versions are the ANSI S3.22-1996 (ANSI 96) and the ANSI S3.22-2003 (ANSI 03). Be sure to check to which standard the hearing aid has been labeled before performing the ANSI test sequence.

2.4.1 Adjusting the hearing aid

In order to test a hearing aid to ANSI specifications, first the hearing aid must be set into a specific test mode. Specifically, the hearing aid should be set to produce the widest possible frequency response range, maximum gain, and maximum output. All special features of the hearing aid such as noise suppression, speech enhancement, or feedback suppression need to be turned off. ANSI 96 specifies that AGC hearing aids should be set to maximum compression. ANSI 03, however, specifies that AGC aids should be set to minimum compression at the beginning of the test. This is to produce a more accurate equivalent input noise (EIN) test.

There are several places in the ANSI test sequence where the analyzer may pause to allow you to make adjustments to the hearing aid. The following adjustments to the hearing aid may be necessary during the ANSI test sequence.

1. Reference Test Position: When the ANSI test sequence is started, the aid should be set to full-on gain. After several measurements have been made, the test sequence may pause, allowing the clinician to adjust the hearing aid to a
reduced reference test position. This is to ensure that the hearing aid will not go into saturation when the rest of the test sequence continues.

2. Telecoil: If the analyzer is set to perform a telecoil measurement, the test sequence will pause to allow the clinician to set the hearing aid to telecoil mode. After the telecoil test is finished, the analyzer will pause again to allow the clinician to set the hearing aid back to normal microphone mode.

3. Compression adjustment: During an ANSI 03 test sequence, the hearing aid analyzer should pause to allow the clinician to set the aid’s compression to maximum. This will occur before the AGC parts of the test: the input/output curves, and attack and release measurements.

**Note:** One of the problems with the ANSI standard is that it specifies a reduced reference test position for many of the measurements, and, in the case of ANSI 03, different compression settings for different parts of the test, but it does not indicate what to do if the aid does not have a volume control or manual compression settings. There are two options: The clinician can use a hearing aid programmer to adjust the gain and compression of the hearing aid as necessary during the test sequence, or the clinician can set the hearing aid for normal use—gain and compression, and run the entire test sequence with those settings. The latter option is more practical, but the test results cannot be expected to exactly match manufacturing specifications. Contact the manufacturer of the hearing aid for more information.

### 2.4.2 Example of ANSI S3.22-1996

*Figure 8* shows an example of a completed ANSI 96 test sequence. The following is a quick explanation of each of the measurements noted on the Figure. A more detailed explanation of each measurement follows.

1. OSPL 90 curve: Full-on gain frequency sweep taken at 90 dB.
2. RESP 50 curve: Reference test gain frequency sweep taken at 50 dB.
3. SPLITS curve: Telecoil response curve
4. MAX OSPL90: Maximum output of the hearing aid and the frequency where it occurs.
5. OSPL90 HFA: High (or special) frequency average of the OSPL90 curve
6. HFA FOG: High (or special) frequency average taken at full-on gain at 50 dB SPL.
7. REFTG Target and Measured: Reference test gain. The Target value is the volume control to aim for when reducing the gain of the hearing aid during the test sequence. The Measured value is the actual measured reference test gain.
8. EQ Inp Noise: Equivalent input noise of the hearing aid.
9. RESP LIMIT: The response limit is determined by taking the three frequency average of the 50 dB response curve and subtracting 20 dB. F1 is the minimum frequency where the RESP 50 curve crosses the response limit. F2 is the maximum frequency where the RESP 50 curve crosses the response limit.
10. THD: total harmonic distortion and the frequencies and amplitudes at which they are taken.

11. HFA-SPLITS: High (or special) frequency average of the telecoil response curve.

12. STS-SPLITS: Difference between the HFA of the microphone curve and the HFA-SPLITS curve.

13. Battery current drain at 1000 Hz at 65 dB SPL.

14. 1/0 Curves: Input/Output response measurements

15. Att & Rel: Measurements of the attack and release characteristics of AGC hearing aids.

2.4.3 OSPL90

The OSPL90 measurement is a pure-tone frequency response using an input level of 90 dB SPL with the gain control of the hearing aid in the full-on position for both AGC and linear hearing aids. OSPL90 measurements are estimates of the maximum possible output of the hearing aid.
Tolerance: none specified

2.4.4 Maximum Output

The maximum output is the maximum amplitude of the OSPL90 curve. The frequency where this amplitude was found is also noted. This measurement is used to determine that the hearing aid will not exceed the patient’s loudness discomfort levels.

Tolerance: \( = \text{OSPL90-MAX} + 3 \text{ dB} \)

2.4.5 OSPL90 HFA (or SPA)

The OSPL90 HFA/SPA is a three frequency average of the OSPL90 curve. The “high frequency average” (HFA) is the average of the amplitudes at 1000, 1600, and 2500 Hz. Those frequencies were chosen because most hearing aids produce “usable” output at those frequencies.

When a hearing aid is not designed to produce usable output levels at all three HFA frequencies, a manufacture may specify three alternate frequencies. These frequencies are referred to as “special purpose average” (SPA) frequencies.

Tolerance: \( \pm 4 \text{ dB} \)

2.4.6 HFA (or SPA) Full-on Gain

The HFA full-on gain measurement is a three frequency average of the hearing aid. In ANSI 96, AGC hearing aids are always tested with a 50 dB SPL input signal, and linear hearing aids are usually tested with a 60 dB SPL input signal. In ANSI 03, all aids are tested at 50 dB SPL.

Although many manufacturers publish complete full-on gain curves in their specs, it is only required that the full-on gain be stated as an HFA or SPA value.

Tolerance: \( \pm 5 \text{ dB} \)

2.4.7 Reference Test Gain

Many of the ANSI S3.22 measurements are performed when the volume control of the hearing aid is set to a reference test gain position. This is to ensure that the hearing aid will not go into saturation during most of the measurements.
Since the average level of conversational speech in quiet is 65 dB SPL, and speech typically varies +12 dB and −18 dB relative to the average level over the course of time, it was determined that the typical maximum peaks of conversational speech occur at 77 dB SPL (65 + 12). Therefore, the ANSI committee determined the hearing aid would not be in saturation if it were tested at 77 dB below the OSPL90 curve.

Many hearing aids do not have a manual volume control. The ANSI standard does not address this problem. To test strictly to ANSI specifications, the clinician should make the necessary adjustments to the hearing aid with a hearing aid programmer. For a faster, more practical test, the clinician may leave the hearing aid set to normal user settings, but the resulting test cannot be expected to exactly match the hearing aid’s specifications.

**Tolerance:** The manufacturer need not state the reference test gain in the specs, but to perform the ANSI measurements correctly, the hearing aid’s volume control should be adjusted to ± 1 dB of the reference test gain for ANSI 96 or ± 1.5 dB for ANSI 03.

### 2.4.8 Frequency Response Curve

The frequency response curve is a pure-tone frequency response using an input level of 50 or 60 dB SPL with the gain control of the hearing aids in the reference test gain position. In ANSI 96, linear hearing aids are tested at 60 dB SPL, and AGC hearing aids are tested at 50 dB SPL. In ANSI 03, all hearing aids are tested at 60 dB SPL.

**Tolerance:** ± 4 dB below 2000 Hz  
± 6 dB above 2000 Hz

### 2.4.9 Frequency Limits

The frequency limit is the frequency range that the manufacturer is required to specify the frequency response curve. The absolute limit is 200 Hz to 5000 Hz. If the frequency limit values f1 and f2 are higher and lower than the absolute values, respectively, then the frequency range is from f1 to f2.

To calculate the frequency limits:

1. Determine the HFA (or SPA) of the frequency response curve.
2. Subtract 20 dB to find the response limit level.
3. Draw a horizontal line across the frequency response graph at the response limit level.
4. Find where the horizontal line crosses the frequency response graph. The lowest and the highest frequency of these crossings is f1 and f2, respectively.
**Tolerance**: none

### 2.4.10 Total Harmonic Distortion

Total harmonic distortion (THD) is a measure of how much a hearing aid distorts a signal that it is amplifying. Whenever a pure-tone signal is amplified, harmonics of this signal are created. THD is the percentage of these harmonics as compared to the total signal produced by the hearing aid. It is taken with the hearing aid set to the reference test gain position.

For hearing aids that use the HFA frequencies, THD is measured at:
- 500 Hz at 70 dB SPL
- 800 Hz at 70 dB SPL
- 1600 Hz at 65 dB SPL

For hearing aids that use SPA frequencies, THD is measured at frequencies half those of the SPA frequencies. For example, if the SPA frequencies are 1600, 2500, and 4000 Hz, THD is measured at 800, 1250, and 2000 Hz.

**12 dB Rule**: If the second harmonic of a pure-tone signal is 12 dB or higher than the fundamental frequency, then the THD does not have to be measured. The reasoning behind this rule is that when a hearing aid amplifies the second harmonic much more than the first harmonic, the large difference in amplification can magnify, to a disturbingly high level, what otherwise would be a moderate THD value.

**Tolerance**: = THD + 3%

### 2.4.11 Equivalent Input Noise

Equivalent input noise (EIN) is a measurement of the internal noise of the hearing aid. It is the difference between a measurement taken with a 60 dB SPL signal (ANSI 96) or a 50 dB SPL signal (ANSI 03), and a measurement taken without an input signal. Both measurements are taken with the hearing aid set to the reference test gain position.

The EIN test was designed with the assumption that the hearing aid being measured has completely linear amplification under the level of the test signal, 50 or 60 dB SPL. If the hearing aid has a compression knee point below this level, the resulting EIN will be artificially high.

**Tolerance**: = EIN + 3 dB
2.4.12 Battery Current Drain

Battery current drain is measured with the use of a battery simulation device that fits into the battery drawer of the hearing aid and plugs into the hearing aid analyzer. This measurement is taken with the hearing aid set to the reference test gain position and the input signal set to 1000 Hz at 65 dB SPL.

_Tolerance_: +20%

2.4.13 Telecoil

The telecoil measurement uses a device known as the telephone magnetic field simulator (TMFS) designed to simulate the magnetic field created by a telephone receiver. When performing this measurement, the hearing aid is set to the reference test gain position and positioned next to the TMFS as it would be positioned next to a hearing aid receiver. That is, the body of a BTE or the faceplate of an ITE/Canal/CIC instrument is placed parallel to the surface of the TMFS. Make sure the hearing aid is in an upright positioning, as it would be placed in the ear, in order to get the best possible test results. An entire frequency sweep is measured. The resulting frequency response is known as the sound pressure level inductive telephone simulator (SPLITS) curve.

_Tolerance_: none

2.4.14 HFA (or SPA) SPLITS

This is the three-frequency average of the SPLITS curve.

_Tolerance_: ± 6 dB

2.4.15 Simulated Telecoil Sensitivity

The STS, as it is known in ANSI 96, or RSETS as it is known in ANSI 03, is the difference between the HFA of the microphone response curve and the HFA of the SPLITS curve. This measurement gives an estimate of how much the patient will have to change the volume control of his hearing aid when switching to its telecoil function in order to get the same overall output.

_Tolerance_: none
2.4.16 Input/Output

An input/output (I/O) measurement characterizes the compression present in the hearing aid. It is tested on AGC hearing aids only. With the hearing aid set to the reference test position and maximum compression, an input frequency is presented at varying input levels from 50 to 90 dB SPL in 5 dB steps. The output level of each signal is measured and plotted against the input level, creating an I/O curve. This measurement is usually done at 2000 Hz, but the manufacturer can specify up to five different I/O curves at 250, 500, 1000, 2000, and 4000 Hz.

**Tolerance:** ± 5 dB

2.4.17 Attack & Release

Attack and release are measurements of how an AGC hearing aid adapts to large fluctuations of amplitude in the input signal. In this measurement, a 55 dB SPL signal is presented to the hearing aid followed by a 90 dB SPL signal, and the “attack” is measured. A second 55 dB SPL signal is presented, and the “release” is measured. This test is usually done at 2000 Hz, but it can also be done at 250, 500, 1000, and 4000 Hz.

**Tolerance:** ± 5 msec or ± 50% (whichever is larger)

2.4.18 Comparison of ANSI 96 and ANSI 03

Here is a chart of the major differences between ANSI 96 and ANSI 03.

<table>
<thead>
<tr>
<th>1996</th>
<th>2003</th>
</tr>
</thead>
<tbody>
<tr>
<td>AGC aids should have their compression set to have maximum effect for all measurements</td>
<td>AGC aids should have their compression set to have minimum effect for all tests except for I/O and attack &amp; release measurements. For those tests, the compression should be set to have maximum effect.</td>
</tr>
<tr>
<td>Measured reference test gain must be within 1 dB of the reference test setting target.</td>
<td>Measured reference test gain must be within 1.5 dB of the reference test setting target.</td>
</tr>
<tr>
<td>Frequency response curve measured at 60 dB SPL for linear aids and 50 dB SPL for AGC aids.</td>
<td>Frequency response curve measured at 60 dB SPL for all aids.</td>
</tr>
<tr>
<td>Equivalent Input Noise (EIN) formula uses an input value of 60 dB.</td>
<td>EIN formula uses an input value of 50 dB.</td>
</tr>
</tbody>
</table>
Full-on gain measured at 60 or 50 dB SPL for linear aids, and 50 dB SPL for AGC aids.

Simulated Telephone Sensitivity (STS)—the difference between the microphone response curve HFA and the telecoil response curve HFA.

| Full-on gain measured at 60 or 50 dB SPL for linear aids, and 50 dB SPL for AGC aids. | Full-on gain always measured at 50 dB SPL. |
| Simulated Telephone Sensitivity (STS)—the difference between the microphone response curve HFA and the telecoil response curve HFA. | Same measurement, but it is now called Relative Simulated Equivalent Telephone Sensitivity (RSETS). |

### 2.5 ANSI S3.42

The ANSI S3.42-1992 (ANSI 92) standard is an optional standard for testing non-linear hearing aids. Unlike the ANSI S3.22 standard that uses pure-tone signals, the ANSI 92 standard uses a broadband noise signal. This means that instead of presenting one frequency at a time like a pure-tone sweep, a broadband signal presents many different frequencies simultaneously. See Section 4.1 for an explanation of the speech weighting used in broadband signals and its affect on measurement results.

One of the reasons that a speech-weighted broadband signal was chosen for the ANSI 92 standard is that it produces more realistic test results than is produced by a pure-tone sweep. When testing any kind of hearing aid with automatic compression, the pure-tone sweep can trigger the compression circuits of the hearing to respond with more amplification in the low frequencies than it would normally occur with a speech input. This is called artificial blooming of the low frequencies. A broadband test signal will not typically cause this test artifact.

![Figure 9: This figure compares the frequency response of an AGC hearing aid to a pure-tone signal (Curve 2), and a composite signal (Curve 1). The pure-tone signal produces artificial blooming of the low frequencies.](image)
See Figure 9 for a comparison of a hearing aid with wideband compression tested with a pure-tone sweep and with the more realistic ANSI S3.42-1992 test signal.

The ANSI S3.42 standard (shown in Figure 10) specifies the following measurements.

1. NSPL90: Average energy (RMS) of a 90 dB SPL noise signal with the hearing aid set to full-on gain.
2. Full-on noise gain: RMS of a 60 dB SPL noise signal with the hearing aid set to full-on gain.
3. Family of gain curves: With the hearing aid set to reference test gain, a series of five frequency response curves are measured using the ANSI S3.42 test signal. The amplitude of the input signal varies from 50 to 90 dB SPL in 10 dB steps.
4. I/O curve: Input/output measurement using the ANSI S3.42 signal with the hearing aid set to reference test gain.

Figure 10: An example of test results according to the ANSI S3.42-1992 standard. This is an optional standard for non-linear hearing aids.
A quick examination of ANSI S3.42 test results can immediately identify the compression characteristics of the hearing aid. If the five frequency response curves produce the same amount of gain (i.e., the curves are placed on top of each other), the hearing aid is amplifying linearly. If the gain of the frequency response curves decreases as the amplitude of the input signal increases, the hearing aid is compressing the signal. The compression knee point can be determined by looking at the I/O curve.
3.0 Real-ear measurements

The frequency response of the patient's ear canal, including any amplification produced by the hearing aid, is tested with the use of real-ear measurements (REMs). While coupler measurements rely on the coupler to represent the average ear canal, REMs use the patient's actual ear. This can be extremely important because the physiology of many ears can differ vastly from a standard 2-cc coupler, causing the response of the hearing aid in the ear to be quite different from the coupler response.

Some potential problems exist with taking REMs. REMs require a certain amount of practice and manual dexterity on the part of the clinician performing the test, and the process of inserting the probe tube and hearing aid can be frustrating for the beginner. Reflective surfaces in the testing environment can cause irregularities in the sound field, and excessive ambient noise can overwhelm the test signal of the real-ear analyzer. These problems can lead to inaccurate or unrepeatable results, defeating the purpose of the measurement. REMs can also be difficult to perform on some patients, especially infants and young children who have a hard time sitting still. A combination of RECD and coupler measurements can be used in place of REMs for these patients. However, armed with an understanding of these potential pitfalls, the clinician can quickly produce accurate and repeatable REMs.

3.1 Real-ear Setup

In order to perform accurate, repeatable REMs, it is critical to have a good test environment. Choose a room that is as large as possible with an area that can be used for testing that is free of vertical and horizontal reflective surfaces such as desks, chairs, and other office furniture. Reflective surfaces should be at least twice as far away from the patient as the sound field speaker will be. It also helps to have a sound treated ceiling, carpeted floor, and drapes on the walls of the room. Small audiological test booths, even if they are sound treated, are not great real-ear testing environments.

Set up the real-ear analyzer so that the speaker is pointed away from any nearby walls or corners. The clinician performing the measurement should be positioned so that he can easily operate the real-ear analyzer and yet be out of the path of the sound field speaker.

Position the patient a distance of 12 to 30 inches from the sound field speaker, as recommended by the manufacturer of the real-ear analyzer. If the patient is positioned too far from the speaker, the real-ear analyzer may not be able to produce a signal at the patient’s ear as loud as the clinician may desire (e.g. 90 dB SPL). If the patient is positioned too close to the speaker, the sound field may contain some speaker irregularities.
Whether the speaker should be positioned at a 0° azimuth, directly in front of the patient, or a 45° azimuth, off to the side of the patient, has been a matter of discussion for some years. Some research has shown that a 45° azimuth positioning produces the most repeatable results. See Figure 11 for an example of a real-ear setup. It is recommended that the clinician decide which angle is most appropriate for the clinical setup available, and use that angle consistently.

### Figure 11

This illustration shows a possible setup for real-ear measurements using 45° azimuth. Other real-ear setups may use 0° azimuth.

### 3.2 Calibration

Before REMs can be performed, the real-ear analyzer microphones should be calibrated and the sound field leveled. REMs typically use two microphones: a reference microphone outside the ear and a probe microphone inside the ear canal. Since every microphone has a slightly different response, in order to ensure accuracy in the REMs, the differences between the two microphones needs to be measured. This allows the real-ear analyzer to compensate for their differences during the REMs. Sometimes this calibration procedure is performed daily as part of the normal use of the analyzers. In other setups, the calibration is stored into the real-ear analyzer’s permanent memory.

Another calibration procedure that needs to be performed before the REM is the process of leveling the sound field speaker. In the leveling process, the real-ear analyzer determines the amount of output required by the sound field speaker in order to generate an input signal of the specified amplitude at the patient’s ear. If the leveling process is not correct, the amplitude of the input signal used in the REM may be incorrect, leading to bad test results. It is recommended to level the real-ear analyzer with the patient in front of the sound field speaker. This will allow the leveling process to account for acoustical irregularities in the sound field that may be generated by the patient’s body and clothes.
See your real-ear analyzer operator’s manual for more details about calibration and leveling.

3.3 Probe tube insertion

All REMs are made with the use of a probe tube connected to the probe microphone. The probe tube is positioned in the patient’s ear canal so as to measure the signal exactly as the patient hears it (Figure 12). In order to minimize errors caused by sound reflections inside the ear, it is important to place the probe tube as close as possible to the eardrum. For most clinical purposes, the end of the probe tube should be within 6 mm from the eardrum. There are several methods in use for inserting the probe tube. Here are two of the most common.

3.3.1 Otoscopic examination

This method involves the use of an otoscope. Many clinicians regard it as the most straightforward way to insert a probe tube. There are three steps:

1. Use an otoscope to examine the ear canal and estimate its length.
2. Insert the probe tube and, using the otoscope, carefully advance it until it reaches within 6 mm of the eardrum. For average adult ears, this will be about 28 mm from the intertragal notch.
3. Mark the probe tube at the intertragal notch. This will indicate whether the probe tube has moved during REM procedure. It can be useful to secure the probe tube in place with surgical tape.

3.3.2 Unaided measurement

This method involves testing the unaided response with the use of a continuous test signal from the real-ear analyzer while inserting the probe tube.

1. Set up the real-ear analyzer for testing, including performing any necessary calibration or leveling procedures. The patient should be placed in front of the loudspeaker with the ear hook on the ear and the probe microphone prepared for insertion.
2. Introduce a continuous broadband signal with the real-ear analyzer.
3. Insert the probe tube carefully, slowly advancing its position inside the ear.
4. Watch the frequency response on the real-ear analyzer display.
As the probe tube is inserted, the response of the ear canal resonance at 6 kHz will decrease as the probe tube encounters the null spot inside the ear canal. Advance the probe tube past this location inside the ear canal until the response at 6 kHz increases and stabilizes. The gain of the frequency response between 6-8 kHz should normally be greater or equal to zero. If the gain at those frequencies is below zero, it could be an indication that the probe tube is not inserted deeply enough into the ear canal.

### 3.4 Prescription Fitting Rules

When performing REMs, the clinician typically adjusts the frequency response of the hearing aid so that it matches a prescription target (or series of targets) to ensure the amplification provided by the hearing aid is appropriate for the patient’s hearing loss.

Linear targets are older prescriptions that are not dependent upon the amplitude of the input signal; the gain prescribed for a 50 dB SPL signal is the same amount of gain prescribed for an 80 dB SPL signal. Non-linear targets prescribe amplification dependent upon the input level of the test signal, typically providing more amplification for quiet signals and less amplification for loud signals, allowing the clinician to compare the hearing aid’s frequency response to targets at different levels for a more comprehensive fitting.

The following are the most commonly encountered real-ear targets:

#### Linear

1. **NAL-RP:** The most common and verified real-ear target used in the hearing healthcare industry. It was developed in Australia by the National Acoustics Lab.
2. **1/2 GAIN:** Samuel Lybarger developed this first fitting rule in the 1940s. His theory was that the gain of the hearing aid should equal one half of the hearing loss.
3. **BERGER:** First described in 1977 and revised as late as 1988, the Berger method was one of the first comprehensive procedures. It was developed by Kenneth Berger of Kent University.
4. **POGO:** An acronym for “Prescription of gain and output,” Pogo was described in 1983 by McCandless and Lyregaard. It is an insertion gain method similar to the half gain rule.
5. **POGO II:** A modification of the POGO fitting rule, POGO II was described by Schwartz, Lyregaard, and Lundh in 1988 for severe hearing losses.
6. **LIBBY:** (1/3 GAIN, 2/3 GAIN): E. Libby described another modification to the POGO method in 1985 and 1986. His theory was that mild to moderate losses should be fit with a gain equal to one third of the loss, moderate losses should be fit with a gain equal to one half of the loss, and severe losses should be fit with a gain equal to two thirds of the loss.
Non-linear

1. DSL-I/O: Developed by the University of Western Ontario, DSL (Desired Sensation Level) was primarily created as a fitting method for children, and later expanded to adults.

2. NAL-NL1: Developed by the National Acoustics Lab in Australia, it expands the original NAL-RP into a method for fitting non-linear hearing aids.

3. FIG6: Developed by Mead Killion of Etymotic Research. It is primarily used in fitting K-AMP hearing aids.

3.5 The Insertion Gain Method

Insertion gain is the most common method used for REMs, although the SPL Method, explained in Section 3.6, is quickly gaining in popularity. During the insertion gain method, the clinician measures the patient’s real-ear unaided gain (REUG) and real-ear aided gain (REAG). The real-ear analyzer calculates the difference between these two measurements to obtain the real-ear insertion gain (REIG). That is, \( \text{REIG} = \text{REAG} - \text{REUG} \). The REIG is the amount of gain that the hearing aid is providing above and beyond the natural resonance of the ear.

The Insertion Gain method involves the following steps, described in more detail in the indicated sections.

1. Set up the patient and real-ear analyzer for testing (Section 3.1).
2. Calibrate and/or level the real-ear analyzer (Section 3.2).
3. Insert the probe tube (Section 3.3).
4. Measure the REUG (Section 3.4.1).
5. Insert the hearing aid without dislodging the probe tube.
6. Measure the REAG (Section 3.4.2).
7. Compare the calculated to the prescribed insertion gain target and make any necessary adjustments to the hearing aid in order for REIG to match the prescribed target.
8. Repeat steps 6-7 as necessary.

Note: The REAG is sometimes confused with the REIG. The REAG is the difference between the signal measured by the probe microphone in the ear canal and the signal measured by the reference microphone outside the ear. That is, it is the gain of the signal after it has been amplified by the hearing aid. The REIG is the difference between the REAG and the REUG. That is, the amplification provided by the hearing aid above and beyond the amplification provided by the natural resonance of the ear canal.

3.5.1 REUR/REUG

The real-ear unaided response (REUR) is the frequency response of the unaided ear to a signal produced by the real-ear analyzer. The REUR includes the input signal used by the real-ear analyzer and is displayed as dB SPL. The real-ear unaided gain (REUG)
is the same measurement, but it refers to the gain of the response. That is, it is the difference between the output measured by the probe microphone in the ear canal and the input signal, usually measured by the reference microphone outside the ear. The REUG is displayed on the graph as dB Gain.

To measure the REUG, set up the patient and real-ear analyzer as described in Section 3.1, calibrate and/or level the sound field speaker as described in Section 3.2, insert the probe tube as described in Section 3.3. The patient should be placed in front of the sound field speaker as illustrated in Figure 11 with a probe tube in the unaided ear (Figure 12). Finally, configure the real-ear analyzer to produce a test signal. The author recommends using a broadband signal with amplitude of 65 dB SPL.

An example of a typical REUG is shown in Curve 1 contained in the bottom graph of Figure 13. Note the frequency response is close to zero before 1000 Hz and contains a
resonance peak between 2-4 kHz. The absence of negative gain between 6-8 kHz is an indication that the probe tube has been inserted deeply enough for accurate measurements.

**Note:** Some researchers have questioned the need to measure the REUG. When the hearing aid is placed in the ear, the natural resonance of the ear is altered. Thus, some regard the REUG as containing extraneous information that is unimportant to the final hearing aid fitting. Clinicians following this train of thought may opt to substitute an average unaided response for the measured REUG when performing REMs using the insertion gain method.

### 3.5.2 REAR/REAG

The real-ear aided response (REAR) is the frequency response of the aided ear to a signal produced by the real-ear analyzer. The REAR refers to the dB SPL output response that includes the input signal used with the measurement. The REAG refers to the dB Gain response in which the input signal is subtracted from the output measurement.

To measure the REAG, the clinician should set up the real-ear analyzer and patient as described in Section 3.5.1, and then carefully insert the hearing aid into the ear without moving the probe tube, as illustrated in Figure 14.

If the hearing aid is linear, it is only necessary to measure the REAR at a medium input level (65 dB SPL). If the hearing aid has compression, it is recommended to measure the REAR at soft, medium, and loud input levels (50, 65, and 80 dB SPL). This will allow the clinician to make sure that the hearing aid amplifies soft sounds with more gain than it uses to amplify loud sounds. The bottom graph in Figure 13 contains REAG measurements at 50 (Curve 2), 65 (Curve 3), and 80 dB SPL (Curve 4).

### 3.5.3 REIG

The real-ear insertion gain (REIG) is the difference between the REAG and the REUG. That is, it is the amplification provided by the hearing aid above and beyond that of the unaided response. A positive REIG indicates that the hearing aid is amplifying above the unaided response. That is, the hearing aid is overcoming the occlusion caused by inserting the hearing aid into the ear canal. A negative REIG indicates an insertion loss. That is, the hearing aid is not providing enough amplification to overcome the occlusion of the hearing aid.
The REIG is usually compared to an REIG prescription target in order to adjust the hearing aid to produce amplification suitable for the patient’s hearing loss. See the top graph in Figure 13 for an example of REIG curves measured at 50 (Curve 6), 65 (Curve 7), and 80 dB SPL (Curve 8). The displayed REIG prescription target (Curve A) is adjusted for the source level of 65 dB SPL. In this example, the hearing aid is producing 5-10 dB excess amplification from 500-1500 Hz and from 2500-4000 Hz. According to the prescription target, this hearing aid fitting should be adjusted by 5-10 dB in those channels, and the REAR measurements should be repeated.

It is important for the clinician to remember that the prescription target is merely a guide for the hearing aid fitting. The clinician should usually come within 5-10 dB of the prescribed target, but it is always important to listen to the needs of the patient being fitted. It may also require several adjustments to the hearing aid over time in order for the patient to accept the prescribed amplification.
3.6 The SPL Method

The SPL Method of performing REMs was first popularized in early 1990s with the DSL Fitting method introduced by the University of Western Ontario. In the SPL Method, the REMs, the prescribed target, and the patient’s thresholds and loudness discomfort levels (LDLs) are displayed together on one graph with all data converted to dB SPL. This display, sometimes known as an SPL-o-gram, has the advantage of allowing the clinician to compare the REMs to the dynamic range of the patient and ensure that the REAR at soft speech levels (50 dB SPL) exceed the patient’s thresholds, the REAR at varying levels from 50-80 dB SPL matches the prescribed target(s), and the REAR at 90 dB SPL is less than the patient’s LDLs. Figure 15 contains an example of an SPL-o-gram.

See Section 3.5.2 for instructions on measuring the REAR. The REUR (Section 3.5.1) can also be measured in the SPL Method in order to ensure that the REAR measurements exceed the REUR (this is especially useful when testing patients with mild to moderate losses), but the REUR is not always measured.

In 1992, Etymonic Design trademarked the term Speechmap to refer to this measurement method. As a result, real-ear measurements displayed in an SPL-o-gram are sometimes referred to as Speech mapping.

3.7 Live speech testing

The use of live or pre-recorded speech as the input signal for the REMs has been quickly growing in popularity during the past few years. These live speech measurements are typically performed using the SPL method in an SPL-o-gram. This allows the clinician to compare the hearing aid’s response to the speech signal to the patient’s HTLs and UCLs. Such measurements are sometimes known as Live Speech Mapping or Visible Speech.

Unlike conventional test signals, the amplitudes of the frequencies components of speech are constantly changing. Typically in a live speech measurement, the real-ear analyzer keeps track of the minimum and maximum amplitudes of each frequency component over the time of the test, allowing the clinician to see the entire amplitude range of the frequency response and how that range compares to the patient’s audiometric data. The analyzer sometimes also displays the long-term average of the real-time measurement. This average can be compared to a prescription target. The clinician should keep in mind, however, that the prescription target may have been created assuming a different speech spectrum than the one used in the speech measurement.

In Figure 16, the area below the patient’s HTLs and above the patient’s UCLs are shaded. The area between these regions is the dynamic range of the patient’s hearing where the signal should be audible to the patient. The dark dotted line contains the average response of the live speech measurement. The dark shaded region (bounded by curves 1u and 1l) around that dotted line is the area of standard deviation where most of the hearing aid response to the
signal was measured. The lighter shaded region (bounded by curves 1U and 1L) contains the area between the maximum and minimum response of the hearing aid per frequency over the duration of the test. That is, the area bounded by Curves 1U and 1L contains the entire frequency response range of the hearing aid during the test.

Live speech measurements are most effective as a counseling tool for the patient. The clinician can use them to show the patient how the hearing aid amplifies actual speech. If the voice of a spouse or family member is used as the input signal for the test, the clinician may opt to perform several different measurements: one with the family member standing next to the patient, and one with the family member standing across the room from the patient. These two tests can demonstrate how close the family member needs to be to the patient in order for the average real-ear response to reach above the patient’s HTLs.

Figure 16: The area below the patient’s HTLs and above the patient’s UCLs is shaded. The dark dotted line (1M) shows the average frequency response of the hearing aid over the duration of the test. Curves 1u and 1l bound a dark shaded region consisting of the standard deviation around the average measurement curve. That is, the area where most of the frequency responses of the test was measured. Curves 1U and 1L contain the maximum and minimum response per frequency of the hearing aid during the test. The shaded area between these curves represents the entire frequency response range of the hearing aid during the test. Curves H, M, and L contain the NAL-NL1 targets for 50, 65, and 80 dB SPL.
4.0 Advanced

This section contains information and testing techniques for real-ear and coupler measurements that would be useful for the clinician who has mastered the basic real-ear and coupler tests described in Sections 2 and 3.

4.1 Effects of Speech Weighting

Pure-tone signals are used for performing ANSI S3.22 hearing aid specification measurements. Pure-tone signals offer a stable test that produce results that are repeatable even when testing with different types of hearing aid analyzers. However, pure-tone signals do not offer the most realistic test results. Broadband signals (i.e., signals composed of multiple frequencies that are presented simultaneously) typically produce results that are more like the real-world response of the hearing aid and are therefore used by most real-ear measurements and non-ANSI coupler measurements. Unlike pure-tone signals, broadband signals do not usually produce equal energy across the frequency band. Most broadband signals are instead speech weighted. That is, they contain more energy in the high frequencies than in the low frequencies.

Speech itself is a broadband signal containing many different frequencies. In English and other similar languages, there is overall greater emphasis in the lower frequencies than in the higher frequencies. Although some syllables have lots of energy in the high frequencies, they tend to be very short in duration and therefore are mostly averaged out when looking at the long-term average of speech. When referring to a signal used to test hearing aids, the signal is said to be speech weighted when the lower frequencies have more energy than the high frequencies.

Curve 1 of Figure 17 illustrates the ANSI S3.42-1992 speech weighting, starting at 3 dB down at 900 Hz and continuing with a gradual high frequency roll off of 6 dB per octave. This curve has a total energy of 70 dB SPL. However, the amplitude at 200 Hz is 58 dB SPL, and the amplitude at 4000 Hz is 47 dB SPL. It is common for even experienced clinicians to be confused at how such a signal has average energy much higher than the energy of any particular frequency component. To explain this phenomenon, imagine a single speaker outputting a signal of 70 dB SPL. Now imagine a second speaker outputting an identical signal. The average energy of the two signals is 73 dB SPL. The ANSI S3.42 signal shown Figure 17 is composed of 79 different frequencies. Those frequencies can be thought of as 79 different speakers each emitting a signal. When all the components are added together, the root-mean-square (RMS) of the signal is 70 dB SPL.

The speech weighting of the broadband signal describes how the energy of a signal is distributed across its frequency components. Some types of speech weighting have more energy in the high frequencies than other types. For instance, Curve 2 of Figure 17 illustrates the speech weighting taken from a CD of sounds produced by the International Collegium of Rehabilita-
Hearing Aid Measurements

tive Audiologists (ICRA). This ICRA speech weighting was used in the development of some
digital hearing aids. Compared to the ANSI S3.42 speech weighting, the ICRA speech weight-
ing has much less energy in the high frequencies.

The speech weighting of a broadband test signal can have a significant impact on the hearing
aid compression circuits that are activated during the test. Figure 18 shows a comparison of a
hearing aid tested with an ANSI-weighted composite signal (Curve 1) and an ICRA-weighted
composite signal (Curve 2). Although these signals have equal RMS energy, the ANSI signal
has more energy in the high frequencies. Since the compression circuit of the hearing aid
provides more gain for softer signals, and the hearing aid determined that the ICRA-weighted
test signal was softer in the high frequencies than the ANSI-weighted test signal, the hearing
aid amplified the ICRA test signal with higher gain than it amplified the ANSI-weighted test
signal.

This effect can also be seen when viewing the measurement results in terms of output on a
dB SPL graph as shown in Figure 19. As in Figure 18, Curve 1 was measured with the ANSI-
weighted composite signal and Curve 2 was measured with the ICRA weighted composite signal. The ICRA-weighted input signal has more energy in the low frequencies and less energy in the high-frequencies than the ANSI-weighted signal. The test results of Figure 19 reflect this difference in input levels. However, the greater gain given by the hearing aid to the ICRA-weighted input signal (demonstrated in the Gain graph of Figure 18) provides almost the same output in the high frequencies as the ANSI-weighted signal.

As demonstrated in Figures 18 and 19, it is critical for the clinician to understand the speech weighting characteristics of the broadband signal used during coupler or real-ear measurements in order to correctly interpret test results. Furthermore, the test signals used by a particular hearing aid analyzer manufacturer can often be quite different from the test signals used by a different manufacturer. Therefore, when comparing measurements across different instrument types, it is important to keep in mind the speech weighting used with each test signal.

### 4.2 Digital Hearing Aid Testing

Many digital hearing aids attempt to distinguish noise from speech in the input signal and to amplify only the speech parts of the signal. This feature is usually called something like “noise reduction”, “noise suppression”, or “speech enhancement.” In most cases, the hearing aid does this by monitoring the signal for modulations. Noise is usually a fairly constant signal while speech varies greatly in amplitude. So the hearing aid suppresses the part of the input signal that is constant while amplifying the part of the signal that is modulated.
Unfortunately, traditional hearing aid test signals such as pure-tone sweeps and non-modulated broadband signals are interpreted as noise by these noise suppression circuits. To test hearing aids with noise suppression, hearing aid analyzer manufacturers have had to develop new modulated test signals. There is no standard modulated signal for testing digital hearing aids, so test results may vary between different types of hearing aid analyzers. In general, however, the use of modulated signals should allow the clinician to perform (non-ANSI) coupler multurve and real-ear measurements without turning off the noise suppression functionality on the digital hearing aid. This will give the clinician a better idea of how the hearing aid will actually perform on the patient’s ear when in normal use mode than measurements obtained when these features are turned off.

A comparison of a measurement performed with a modulated signal to one performed with a non-modulated signal is an easy way to test the noise suppression technology on the hearing aid. This allows the clinician to determine exactly how much the hearing aid is suppressing noise. Figure 20 illustrates a noise-suppressing digital hearing aid tested with a non-modulated ANSI-weighted composite signal (Curve 1), and a modulated ANSI-weighted composite signal of the same amplitude (Curve 2). There is a 5-10 dB difference in the output levels of these measurements at nearly all frequencies.

Some hearing aid analyzers can also add a continuous “bias” signal to the modulated signal. This bias signal creates a noise at a particular frequency in an otherwise modulated signal. This can allow the clinician to see how the hearing aid responds to noise at different frequencies and to determine if the noise in one channel affects the response of the aid in its other channels. Figure 21 shows a hearing aid tested with a modulated signal containing noise at 500 Hz (Curve 2), 2000 Hz (Curve 3), and 4000 Hz (Curve 4). The original modulated signal without the bias noise component is shown in Curve 1 for comparison. Notice how the hearing aid suppresses the noise present in one channel without significantly altering the amplification in the other channels. The use of the bias signal can be a useful tool for determining the noise suppression characteristics of the digital hearing aid.

Although the measurements illustrated in Figures 21 are coupler measurements, the same technique is applicable to REM.
4.3 Open ear Hearing Aid Testing

Most CIC, ITE, ITC, and BTE (with an attached earmold) hearing aids occlude the ear when worn. This occlusion can lead to a phenomenon known as the occlusion effect, in which the bone-conducted vibrations of the patient’s own voice is trapped inside the ear, unable to escape through the ear canal. Traditionally, venting has been used to decrease this effect by allowing those vibrations to escape through the vent of the hearing aid. Larger vents allow more sound to escape than smaller vents, thus decreasing the occlusion effect, but also increasing the chance of feedback. Open ear style hearing aids can be thought of as the logical extension of this venting solution to the occlusion effect, but, because of the feedback issue, were not practical until recently with the evolution of advanced feedback suppression technology in digital hearing aids.

When performing real-ear measurements on open ear style hearing aids, the clinician should keep several factors in mind. First, open ear style hearing aids typically provide mild to moderate gain in the mid to high frequencies (1000 Hz to 8000 Hz), but little to no amplification in the low frequencies (below 1000 Hz). However, at the time of the writing, all real-ear targets are designed for more traditional style hearing aids that do provide amplification in the low frequencies. Therefore, the clinician should not expect to match the target in the low frequencies with an open-ear style hearing aid. Second, the feedback suppression technology on the open-ear style hearing aid can sometimes interfere with the reference microphone measurement of the sound field outside the ear. Also, amplified sound from the hearing aid can escape the ear and be measured by the reference microphone as part of the original input signal. This can also affect measurement results.
This means that in some cases, turning the reference microphone off may increase the measured gain of the REAG, and in other cases, turning the reference microphone off may decrease the response of the REAG. In still other cases, the status of the reference microphone may have no effect at all on the real-ear measurements. At this time, the author recommends that the clinician disable the reference microphone of the real-ear analyzer (if possible) when testing open ear style hearing aids.

The reference microphone is normally used by the real-ear analyzer to automatically adjust to slight changes made to the sound field caused by the patient moving. Therefore, when the reference microphone is disabled, the clinician should take special care that the patient remain still and that the testing environment not be disturbed after the sound field leveling process.

### 4.4 Testing Directional Hearing Aids

When directional hearing aids are discussed and programmed, there are usually references to polar plots and patterns that illustrate how the directionality of the hearing aid works. These plots are usually taken from careful measurements made by the manufacturer in anechoic chambers with analysis equipment not available to the average clinician. However, clinical hearing aid analyzers can usually be adapted to take some basic measures of directionality. Typically, this involves taking a frequency response measurement with the aid pointed towards the sound chamber speaker, and another measurement with the aid pointed away from the sound chamber speaker. These responses can be called the “forward” and “reverse” measurements.

These methods will vary according to the setup for the hearing aid analyzer. One such method is explained here.

1. Determine where the sound chamber speaker is located and the direction in which it is pointed.
2. Position the directional hearing aid so that the top microphone is pointed towards the sound chamber speaker. In Figure 22a, the sound chamber speaker is pointed at a 45 degree angle from right to left. By placing the hearing aid pointed towards the right, the directional microphone are pointed towards the speaker.
3. Measure a frequency response measurement using an input signal of 50 dB SPL. The amplitude of the test signal should be lower than the compression knee point of the hearing aid so that it is not in compression during the directional test.

![Figure 22a](image-url): This is an example of a setup for the “forward” response of a directional test. The speaker in the sound chamber is situated at an angle in the bottom of the chamber from right to left. Therefore, this hearing aid is pointed towards the speaker.
4. Position the aid so that the top microphone is pointed away (usually about 135°) from the sound chamber speaker. See Figure 22b.

5. Measure a frequency response using the same input signal that was used in step 3 (50 dB SPL).

6. Compare the two measurement curves. The reverse measurement should be lower than the forward measurement, giving the hearing aid directional advantage for signals in front of the aid. See Figure 23.

Real-ear measurements can also be used to measure the hearing aid’s front to back directionality. The concept for the real-ear test is the same as the coupler test explained above. That is, one measurement is performed with the sound field speaker placed in front of the patient (the “forward” measurement), and one measurement is performed with the sound field speaker placed behind the patient (the “reverse” measurement). The insertion gain test mode on the real-ear analyzer is a particularly useful tool for testing directionality because it subtracts one curve from another curve. When measuring directionality, the reverse measurement can be subtracted from the forward measurement to show how much more amplification the hearing aid provides for signal in front of the patient than for signals behind him.

The following are basic instructions for performing a real-ear directionality test:

1. Set up the patient and the real-ear analyzer for REMs. Place the sound field speaker behind the patient at 180° or 135° azimuth. Make sure to calibrate and/or level the sound field speaker. See Figure 24.
2. Configure the real-ear analyzer to perform an “unaided” measurement. However, perform this test with the hearing aid on the ear and the sound field behind the patient. Use a 50 dB SPL broadband modulated signal, if possible. The “unaided” measurement as shown on the real-ear analyzer screen is actually the “reverse” directional measurement.

3. Move the sound field speaker so that it is placed in front of the patient at 0° or 45° azimuth in the usual REM configuration. Relevel the sound field speaker if it is possible to do so without deleting the reverse measurement on the real-ear analyzer screen.

4. Configure the real-ear analyzer to perform an “aided” measurement. Use the same signal type and amplitude from Step 2 (50 dB SPL modulated broadband signal, if possible). The “aided” measurement as shown on the real-ear analyzer screen is actually the “forward” directional measurement.

5. Look at the resulting “insertion gain” curve on the analyzer. This is the difference between the reverse and forward directional measurements. That is, the “insertion gain” curve shows the directional advantage of the hearing aid as a function of frequency.

Figure 25 illustrates a reverse measurement (Curve 1) using the unaided curve slot on the real-ear analyzer, a forward measurement (Curve 2) using the aided curve slot, and the directional advantage of the hearing aid (Curve 6).
4.5 Problems with Group delay

Digital hearing aids have advanced the field of amplification significantly by providing flexible amplification with low distortion and many features not possible with analog technology. However, one of the properties of digital technology not always mentioned in sales literature is that it always takes time to process sound through a digital hearing aid. Imagine the digital hearing aid as a miniature computer: it takes an analog sound, turns it into digital data, performs some kind of algorithm to amplify the sound, and turns the digital data back into an analog sound wave for the ear to hear. The amount of time that it takes the digital hearing aid to perform this process is known as the digital processing delay. The term group delay technically refers only to the amount of time it takes for the hearing aid to perform the digital algorithm on the signal, but it is also often used interchangeably with digital processing delay.

Group delay is particularly a problem in open ear fittings or in hearing aids with large vents. Sound can travel to the unaided ear faster than through the hearing aid, creating an echoing effect. In monaural fittings, this effect may cause localization problems. Unfortunately, most research into hearing aid group delay has been performed with binaural occluded fittings. Although it has been shown that the group delay present on most digital hearing aids is not significant enough to cause problems in binaural occluded fittings, there has not been enough research done to determine how much group delay is acceptable for fittings with large vents, open ear style hearing aids, or monaural losses. It is the opinion of the author that most
people seem to be able to accept a group delay of 5-6 milliseconds, although that amount of delay may be unacceptable for some people. If the clinician wants to be conservative, a group delay of no more than 3 milliseconds should be acceptable for all patients.

A test has been developed to easily measure group delay with a clinical hearing aid analyzer. In this test, the hearing aid analyzer delivers a short impulse signal to the digital hearing aid and measures the aid’s response. The analyzer determines how much time it takes for the hearing aid to process the impulse and produce a signal. Figure 26 demonstrates the group delay test. Notice how the hearing aid produces a series of impulses in response to the test signal. In this example, the hearing aid analyzer determines the response impulse with the highest amplitude (the maximum impulse). It then takes the first impulse with at least half the amplitude of the maximum impulse (first significant impulse). The delay of the first significant impulse is subtracted from the group delay of the hearing aid analyzer to determine the 5.5 ms group delay of the hearing aid.

### 4.6 Testing Children

Special care and consideration needs to be taken when performing REMs on infants and small children. There are two main issues: first, the size of a child’s ear canal changes quickly until the age of about two years. This affects the types of real-ear targets that are suitable for use when fitting a young child with hearing aids. Second, it may be practically impossible to perform a real-ear sound field measurement on a child who refuses to remain still long enough to set up and perform the real-ear measurement.

The size of the ear canal affects the amplification at the eardrum. A smaller ear canal will produce more amplification given the same amount of energy than a larger ear canal. Therefore, if a certain amount of output (measured at the ear canal) is appropriate for a child, the amount of insertion gain necessary to reach that output would be less for a 2 month old child than it would be for a 24 month old child. That is, even though it may be appropriate to prescribe the same amount of output as measured at the ear canal for a 2 month old as for a 24 month old, the changes in the ear canal as the child ages would require different insertion gain prescriptions for different ages. For this reason, real-ear prescriptions for infants and children are usually given as dB Gain or dB SPL targets instead of the more traditional dB insertion gain target. (It should be remembered that dB Gain is the difference between the output measured by the probe microphone in the ear canal and the input signal normally measured by the reference microphone outside of the ear. Insertion Gain is the difference between the aided gain and the unaided gain.)
It is often difficult to perform an REM on an infant or small child. The process of inserting a probe microphone and a hearing aid in the child’s ear, and having that child sit still long enough to perform an accurate REM can be practically impossible for the clinician. However, since a small child is unable to communicate to the clinician about the hearing aid fitting, it is critical that a prescription target and a way to measure the response of the hearing aid against that target be used in the fitting process. The researchers of the DSL fitting method at the University of Western Ontario have popularized the “coupler method” of fitting hearing aids on children.

Using the “coupler method,” the real-ear prescribed targets are converted into coupler targets suitable for comparison to coupler measurements. The real-ear to coupler difference (RECD, described in Section 4.7) is used in this conversion process to account for the ear canal resonance of the child’s ear, and how it differs from the resonance of the coupler used when adjusting the hearing aid to target. The idea is that, even if an REM is impossible to perform on the child, the clinician may be able to perform the RECD on the child, which is a simpler measurement. If the clinician is unable to measure the RECD, the average RECD based on the child’s age can be substituted to obtain a suitable coupler target.

Sometimes the “coupler method” is referred to a “simulated real-ear.” In the simulated real-ear process, the RECD is used to convert coupler measurements into measurements suitable for comparison against a real-ear target. These converted coupler measurements are referred to as “simulated real-ear.” In other words, when using coupler targets, the real-ear targets are converted for comparison against coupler measurements. In simulated real-ear, the coupler measurements are converted for comparison against real-ear targets.

### 4.7 RECD

The real-ear to coupler difference (RECD) is the difference between the resonance of the ear canal and the resonance of a 2-cc coupler. The RECD is used to convert real-ear prescription targets into “coupler targets,” allowing the clinician to adjust the hearing aid fitting to a prescribed target using a coupler measurement instead of an REM. This process is most commonly used when fitting hearing aids on an infant or small child, but it can also be used when adjusting the hearing aid fitting of an adult.

The RECD consists of two measurements: a coupler measurement and an REM. Both measurements should use the same transducer, input signal, and measurement microphone, or include correction factors for any differences. The coupler measurement is subtracted from the REM to obtain the RECD. Unfortunately, there is no standard method for performing an RECD. Different real-ear analyzer manufacturers use different transducers and different measurement methods, which can lead to different measurement results. Research is still being done to determine the most accurate RECD method. One common method for performing the RECD is described in Sections 4.7.1 and 4.7.2.
4.7.1 Coupler measurement

To perform the coupler measurement portion of the RECD:

1. Attach the probe microphone to the HA-2 coupler using a probe microphone adapter. See Figure 27.
2. Plug an insert earphone into the real-ear analyzer and configure the analyzer to use it as the measurement transducer.
3. Attach the insert earphone to the tubing of the HA-2 coupler.
4. Configure the analyzer to produce a 50 dB SPL pure-tone sweep.
5. Perform the measurement.

4.7.2 Real-ear measurement

To perform the REM portion of the RECD:

1. Insert the probe microphone into the ear in the usual manner. The sound field speaker is not used in this measurement, so it should not be leveled.
2. Plug the insert earphone into the real-ear analyzer and configure the analyzer to use it as the measurement transducer. Make sure to use the same insert earphone that was used for the coupler measurement.
3. Insert the insert earphone into the patient’s ear using a foam ear tip or the patient’s custom earmold. See Figure 28.
4. Configure the analyzer to produce a 50 dB SPL pure-tone sweep.
5. Perform the measurement.

4.7.3 Calculation

To calculate the RECD, subtract the coupler measurement from Section 4.7.1 from the REM from Section 4.7.2. This difference is the RECD.
4.8 REDD

In the real-ear SPL Method described in Section 3.6, the patient’s audiometric information, originally measured in dB HL, is converted to dB SPL. This conversion process uses a transform called the real-ear to dial difference (REDD). Although most real-ear analyzers will use an average REDD to perform this conversion, using a measured REDD will increase the accuracy of the dB HL to dB SPL conversion.

To measure the REDD:

1. Insert the probe microphone into the patient’s ear in the usual manner. The sound field speaker is not used in this measurement, so it should not be leveled.

2. Insert an insert earphone attached to an audiometer into the patient’s ear using a foam ear tip or the patient’s custom earmold as shown in the RECD setup in Figure 28. Alternately, place TDH headphones attached to the audiometer onto the patient’s head.

3. Configure the real-ear analyzer to perform a spectrum analysis measurement using an external signal as the input.

4. Configure the audiometer to produce a 70 dB HL tone into the earphones at 250 Hz.

5. Note the RMS OUT measurement of the real-ear analyzer.

6. Subtract the value obtained in Step 5 from 70. This is the REDD for 250 Hz.

7. Repeat steps 3-6 to obtain the REDD for other audiometric frequencies. As a minimum, measurements should be obtained at 500, 1000, 2000, and 4000 Hz.
5.0 Conclusion

The past decade has brought incredible advancements in hearing aid technology including the widespread use of digital technology, noise suppression, feedback cancellation, and open ear fitting. As hearing aids have grown more sophisticated, so have the measurements needed to verify their functionality. Improvements in testing technology have been made, including modulated signals to ensure accurate measurements of noise suppressing hearing aids, convenient tests of directionality, and techniques for testing open fit hearing aids.

Advances in hearing aid technology have also made it even more critical to perform measurements of the basic characteristics of hearing aids including frequency response, compression, distortion, and the appropriateness of the amplification for the hearing loss of the patient. Most of the tests and tools described in this chapter can be performed quickly and easily by the clinician, especially when included as part of the daily office routine.
Hearing Aid Measurements
6.0 List of Figures

Figure 1: The measurement microphone is inserted into the 2-cc coupler, which is connected to the hearing aid. A battery pill is used to measure the battery current of the hearing aid.

Figure 2: The HA-1 2-cc direct access coupler is used for testing canal-type hearing aids (ITE, ITC, CIC).

Figure 3a: Roll the putty into a rod
Figure 3b: Attach the putty to the hearing aid.
Figure 3c: Make sure the sound opening of the hearing aid is unobstructed.
Figure 3d: Cover the vent of the hearing aid with putty.
Figure 3e: Seal the hearing aid to the coupler so that there are no leaks

Figure 4: The HA-2 2-cc coupler is used for testing BTE hearing aids.

Figure 5: The Zwislocki is used to produce a real-ear like frequency response.

Figure 6: The CIC coupler is a non-standard coupler used to produce a real-ear like frequency response of a CIC hearing aid

Figure 7: This figure shows a comparison of a frequency response of a CIC hearing aid using an HA-1 coupler (Curve 1), and a CIC coupler (Curve 2).

Figure 8: This figure shows an example of a hearing aid tested to the ANSI S3.22-1996 standard.

Figure 9: This figure compares the frequency response of an AGC hearing aid to a pure-tone signal (Curve 2), and a composite signal (Curve 1). The pure-tone signal produces artificial blooming of the low frequencies.

Figure 10: An example of test results according to the ANSI S3.42-1992 standard. This is an optional standard for non-linear hearing aids.

Figure 11: This illustration shows a possible setup for real-ear measurements using 45° azimuth. Other real-ear setups may use 0° azimuth.
Figure 12: This figure illustrates a probe tube inserted into the unaided ear. In this example, the reference microphone is separated from the probe microphone and placed above the ear.

Figure 13: This figure is an example of an insertion gain measurement. One unaided curve has been measured (Curve 1) using a composite input signal of 65 dB SPL. Four aided curves have been measured (Curves 2, 3, and 4) using a Digital Speech input signals of 50, 65, and 80 dB SPL, respectively. Curves 6, 7, and 8 show the insertion gain curves corresponding to aided curves 2, 3, and 4 subtracted from unaided curve 1. Curve A in the upper graph is the NAL-NL1 prescription target for 65 dB SPL and can be directly compared to Curve 7.

Figure 14: This figure illustrates an example of a real-ear aided measurement setup. It is important not to dislodge the probe tube when inserting the hearing aid into the ear.

Figure 15: This screenshot shows an example of an SPL-o-gram. Curve T contains the patient’s HTLs converted to dB SPL. Curve 1 in an REUR. Curves 2, 3, and 4 are REAR measurements at 50, 65, and 80 dB SPL, respectively. Curves L, M, and H contain the NAL-NL1 prescription targets for the REAR measurement curves. The dotted line that runs just off the top of the graph contains the patient’s LDLs.

Figure 16: The area below the patient’s HTLs and above the patient’s UCLs is shaded. The dark dotted line (1M) shows the average frequency response of the hearing aid over the duration of the test. Curves 1u and 1l bound a dark shaded region consisting of the standard deviation around the average measurement curve. That is, the area where most of the frequency responses of the test was measured. Curves 1U and 1L contain the maximum and minimum response per frequency of the hearing aid during the test. The shaded area between these curves represents the entire frequency response range of the hearing aid during the test. Curves H, M, and L contain the NAL-NL1 targets for 50, 65, and 80 dB SPL.

Figure 17: Curve 1 in this figure shows the ANSI S3.42 speech weighting. Curve 2 shows the ICRA speech weighting. The overall energy of each curves is 70 dB SPL.

Figure 18: Curve 1 shows the gain of a hearing aid measured with an ANSI S3.42-weighted composite signal. Curve 2 shows the gain of the same aid measured with an ICRA-weighted composite signal. Because the ICRA-weighted signal contains less energy in the high frequencies, it is amplified more by the hearing aid than the ANSI-weighted signal.

Figure 19: This figure shows the gain measurements of Figure 18 converted to SPL. Curve 1 was measured with an ANSI-S3.42 weighted test signal. Curve 2 was measured with an ICRA-weighted test signal. Although the ICRA signal contains less energy in the high frequencies than the ANSI-weighted signal, the overall output is nearly the same between the two signals because the hearing aid amplifies the lower amplitudes of the ICRA high frequencies more than it amplifies the same frequencies in the ANSI-weighted signal.
Figure 20: This graph contains two REAR measurements. Curve 2 was measured using a modulated test signal. Curve 3 was measured using a continuous test signal. This comparison demonstrates the noise suppression feature of the hearing aid when exposed to continuous noise.

Figure 21: This graph shows the effect of noise on the frequency response of the hearing aid in different channels of the hearing aid. Curve 1 was measured with a modulated signal with no noise. Curves 2, 3, and 4 used a modulated input signal containing a bias at 500, 2000, and 4000 Hz, respectively. Notice how the bias signal lowered the gain of the frequency response in one channel of the hearing aid, but maintained the amplification in the other channels.

Figure 22a: This is an example of a setup for the “forward” response of a directional test. The speaker in the sound chamber is situated at an angle in the bottom of the chamber from right to left. Therefore, this hearing aid is pointed towards the speaker.

Figure 22b: This is an example of a setup for the “reverse” response of a directional test. The speaker in the sound chamber is situated at an angle in the bottom of the chamber from right to left. Therefore, this hearing aid is pointed away from the speaker.

Figure 23: Curve 1 shows the coupler “forward” response of a directional hearing aid. Curve 2 shows the “reverse” response of the hearing aid using the same input signal. This comparison clearly demonstrates the directional advantage of the hearing aid.

Figure 24: This is an example of a real-ear “reverse” setup for a directional microphone. The speaker in this picture is at 180° azimuth.

Figure 25: This figure shows a directional real-ear measurement. Curve 1 is a measurement of the “reverse” response of the hearing aid. Curve 2 shows the “forward” response. Curve 6 in the top graph demonstrates the difference between the forward response and the reverse response. That is, the directional advantage of the hearing aid.

Figure 26: Example of a group delay measurement of a digital hearing aid. The system delay of the hearing aid analyzer is shown as a dotted line at 0.5 ms. The system delay of the hearing aid is shown as the dotted line at 6 ms. The delay of the hearing aid is calculated as 5.5 ms.

Figure 27: This figure contains an example of a setup for the coupler portion of an RECD measurement. An adapter is used to insert the probe microphone into the coupler. An insert earphone is used to drive the input signal through the tubing of the coupler.

Figure 28: This figure illustrates an example of a setup for the real-ear portion of an RECD measurement. The probe tube is inserted into the ear as normal. An insert earphone is used to drive the input signal.
7.0 Bibliography


