Scientific Foundations of Audiology

Perspectives from Physics, Biology, Modeling, and Medicine

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CHAPTER 1

Middle-Ear Reflectance: Concepts and Clinical Applications

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The middle ear is a complex sound transmission system that converts airborne sound into cochlear fluid-born sound, in a relatively efficient way, over the bandwidth of hearing (about 0.1–15 kHz). The middle ear is the gateway to the auditory system, and it is involved in nearly every audiologic test. It is therefore critical to assess middle-ear status in any audiologic evaluation and, in the case of abnormal middle-ear function, pinpoint the source of pathology to enable an appropriate medical intervention. By the use of wideband acoustic measurements, the middle-ear structures can be noninvasively probed across the wide frequency range of hearing, allowing clinicians to make nuanced interpretations of hearing health. The term *wideband acoustic immittance* (WAI) has recently been coined as an umbrella term to identify a variety of acoustic quantities measured in the ear canal (Feeney et al., 2013). Here we focus primarily on wideband reflectance, from which other WAI quantities may be derived. The *reflectance* is defined as the ratio of reflected to forward pressure waves.

A middle-ear reflectance measurement involves inserting an acoustic measurement probe into the ear canal, fitted with an ear tip designed to create a sealed ear-canal cavity (Figure 1–1). A hearing aid loudspeaker in the probe transmits wideband sound into the ear canal. Any reflected sound, related to structures of the middle ear, is measured by the probe microphone. This probe is calibrated in such a way that the absorbed and reflected pressures in a cavity may be determined.

Reflectance measurements are clinically practical to make: The measurement takes less than a minute and the ear does not require pressurization. The

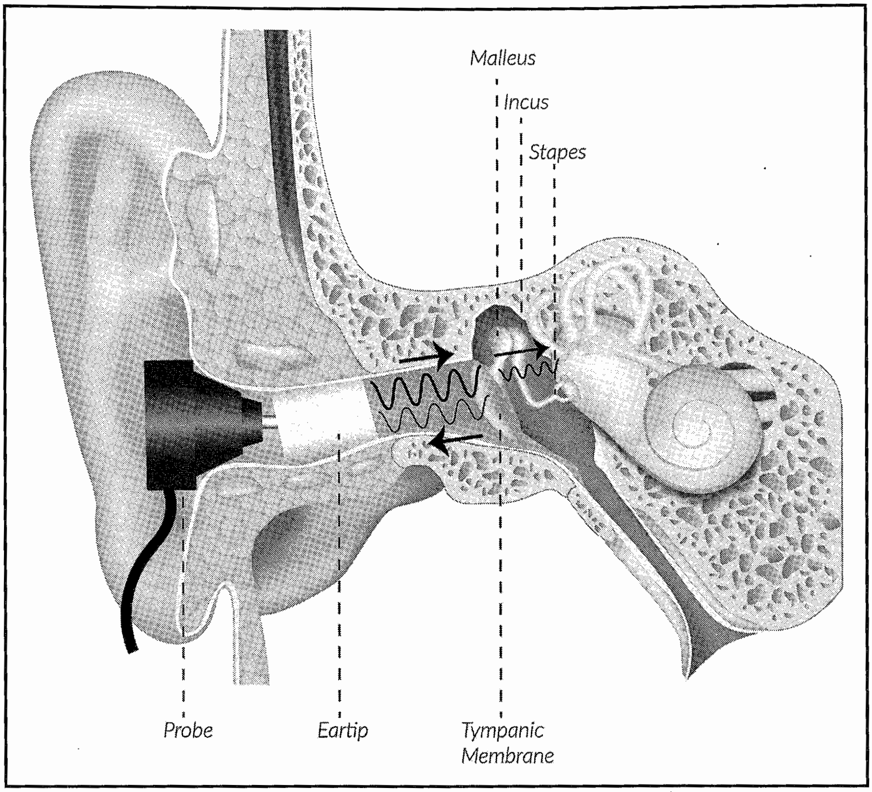

Figure 1–1. Probe configuration in the ear canal to measure middle-ear reflectance, showing the acoustic signal traveling down the ear canal until it reaches the TM. At the TM, the sound is partially reflected back into the ear canal and partially absorbed into the middle ear.

same probe can be used for other audiologic tests, such as otoacoustic emission (OAE) tests and pure-tone hearing threshold testing. Such testing, when a microphone is used in the ear canal, is known as *real-ear* testing. Given knowledge of the reflectance, it is possible to correct for troublesome ear canal standing waves, which can produce large artifacts in the real-ear calibrations. Alone, or together with other audiologic measurements, middle-ear reflectance measurements can help identify many abnormal conditions which may lead to conductive hearing loss (CHL), including degrees of otitis media, tympanic membrane (TM) perforations, otosclerosis, and ossicular disarticulation. The method is noninvasive, fast, and clinically available.

In this chapter, we cover the theoretical principles of middle-ear reflectance. We then move to clinical applications, showing how normal middle ears behave and how abnormal middle ears differ. We offer advice on how to make quality measurements and provide suggestions for future research.

Background to Middle-Ear Assessment

Noninvasive assessment of middle-ear status is of great importance in hearing health care. An early approach to middle-ear assessment is that of *tympanometry* (e.g., Feldman, 1976; Shanks, 1988), and it is still the clinical gold standard. The method relies on measurements at low frequencies (e.g., probe tones at 226 Hz and 1,000 Hz are commonly used) and provides no information on the status of the middle ear at higher frequencies relevant to speech perception (e.g., 0.2–8.0 kHz). The methods employed in tympanometry were developed prior to the introduction of digital technology, and these methods reflect the limitations of that era.

Reflectance of sound from the TM and the acoustic impedance of the middle ear are different facets of the same underlying mechanism. Historically, acoustic impedance of the ear was the first to be measured and studied (West, 1928). There is a substantial body of research on the acoustic impedance of the ear. Metz (1946) developed the first clinical instrument for measuring the acoustic impedance of the ear. This instrument was not easy to use and clinical measurement of acoustic impedance proceeded at a slow pace until more practical instruments were developed (Møller, 1960; Terkildsen & Nielsen, 1960; Zwislocki & Feldman, 1970). Tympanometry, the measurement of the middle-ear acoustic impedance as a function of static pressure in the ear canal, provided useful clinical data. Thus, practical instruments were developed for measurements of this type. The 1970s saw a rapid growth in the use of tympanometry, which is widely used today in audiologic evaluations (Jerger, 1970).

The introduction of small, inexpensive computers in the mid-1980s paved the way for a new generation of digital test equipment with capabilities well beyond that of conventional electronic instrumentation. It also facilitated new ways of thinking about audiologic measurement, resulting in the development of innovative wideband techniques. The evolution of wideband reflectance measurement allows for more detailed diagnostic assessment of the middle-ear status than the previous approach based on tympanometry. Early reflectance studies were conducted by Keefe, Ling, and Bulen (1992); Keefe, Bulen, Arehart, and Burns (1993); and Voss and Allen (1994).

The use of reflectance measurements in a computer-based system does not preclude the use of acoustic impedance data, where appropriate. Acoustic reflectance and acoustic impedance are both WAI quantities; different facets of the same underlying mechanism. If one is known, the other can be computed by means of a mathematical transformation. This mathematical transformation can be implemented conveniently in a computer-based instrument.

Acoustics of the Outer and Middle Ear

When a sound wave travels down the ear canal toward the TM, the acoustic power is continuous until it reaches an *impedance discontinuity*, such as the

Propagation of Sound: The Basics

Many of the concepts in WAI, including reflectance, are defined in mathematical or physics terms. This creates a problem for clinicians and others without the necessary background. Here we explain some acoustical concepts in lay terms.

The transmission of sound in the ear canal can be approximated quite well by a tube with a fixed diameter equal to that of the average adult ear canal. The tube is terminated at one end by a loudspeaker that delivers an acoustic signal in the frequency range up to at least 10,000 Hz. One may imagine that the air in the tube is partitioned into a very large number of infinitesimally thin discs (Beranek, 1949); each disc can be thought of as consisting of a layer of air particles. These discs of air are compressed or expanded by an applied force, such as a change in air pressure (air molecules will spread out from an area of high pressure to an area of lower pressure), and will return to their original volume once the applied force is removed.

Consider now what happens when the loudspeaker at one end of the tube generates an acoustic signal. When the speaker diaphragm moves inward, it displaces and compresses the adjacent discs of air, which then displace and compress the next layer of air, and so on. By this means, the in and out movements of the transducer diaphragm create a pressure wave that travels down the tube at the speed of sound, about 343 m/s at 20°C. The velocity of each disc of air about its quiescent position (the position of the disc at rest) multiplied by its cross-sectional area is known as the *volume velocity*, as the product of velocity and cross-sectional area encompasses a moving volume.

The air in the tube opposes being displaced and compressed by the transducer diaphragm. The force exerted by the transducer diaphragm is equal to the pressure times the area of the diaphragm. The *work* done by the force is equal to force times the displacement, and is stored as energy in the air as it travels along the tube. The *acoustic power*, $P(f)$ (the force times the volume velocity, often expressed in watts), inserted into the tube is equal to the rate of work done. The power propagated down the tube is transmitted without significant loss through the tube via the air.

TM. Impedance discontinuities result in frequency-dependent reflections of the sound wave, which we quantify using wideband reflectance.

The acoustic variables discussed in this section may be defined either in the time or frequency domain. It is important to always be aware of which domain is under consideration. In this chapter, we work almost exclusively in the frequency domain, where all variables are functions of frequency, f. These variables are also a function of location. For measurements in the ear canal, we define $x = 0$ as the measurement probe location and $x = L$ the TM location.

Pressure and Volume Velocity Waves

We denote the forward traveling pressure wave as $P_+(f,x)$ [Pa], using the plus sign subscript to signify the forward direction (toward the TM). This wave is a function of both frequency f (in Hz) and location and has units of Pascals. Similarly, the reflected, backward traveling *retrograde* pressure wave is denoted $P_-(f,x)$. At any location in the ear canal, the total pressure $P(f,x)$ is defined as

$$P(f,x) = P_+(f,x) + P_-(f,x). \tag{1}$$

The pressure is a scalar quantity (it has no direction). Any change in the pressure results in a force, which is a vector quantity (it has direction); this force leads to the motion (velocity) of air molecules in the direction of the force.

The corresponding acoustic *volume velocity* $U(f,x)$ may be decomposed into forward $U_+(f,x)$ and reverse $U_-(f,x)$ traveling portions, as

$$U(f,x) = U_+(f,x) - U_-(f,x). \tag{2}$$

The volume velocity is a vector quantity, which accounts for the change in sign of Equation 2 (here positive U_- values indicate propagation of the retrograde wave toward the probe, and positive U_+ values indicate propagation of the forward wave toward the TM).

The *complex acoustic reflectance*, which we represent using the uppercase Greek letter "Gamma," is defined as the ratio of retrograde to forward traveling pressure (or velocity) waves

$$\Gamma(f,x) = \frac{P_-(f,x)}{P_+(f,x)} = \frac{U_-(f,x)}{U_+(f,x)}. \tag{3}$$

Since $\Gamma(f,x)$ is complex, it may be expressed either as the sum of real and imaginary parts, or in terms of a magnitude and phase. The utility of the complex reflectance (as compared to other WAI quantities, such as impedance and admittance) is that the acoustic power is proportional to the square of the pressure. Thus, the squared magnitude of the reflectance describes the ratio of reflected to incident power (a value ranging between 0 and 1) as a function of frequency, while the reflectance phase codifies the latency of the reflected power (e.g., the depth at which the reflection occurs). Additionally, power absorbed by ear (potentially including the ear canal, middle ear, and inner ear) may be quantified as one minus the ratio of power reflected. The *power reflectance* at the probe may be defined as $|\Gamma(f,0)|^2$; thus, the power absorbed by the ear is $1 - |\Gamma(f,0)|^2$. These properties of reflectance are more intuitive than impedance for formulating diagnoses of middle-ear pathologies.

For reference, the *complex acoustic impedance* is defined as the total pressure over the total volume velocity

$$Z(f,x) = \frac{P(f,x)}{U(f,x)}. \tag{4}$$

The *complex acoustic admittance* is given by $Y(f,x) = \frac{1}{Z(f,x)}$ and various other WAI quantities may be calculated from $Z(f,x)$ and $Y(f,x)$, as outlined in Appendix 1–A. This variety of immittance quantities can be confusing, so it is important to remember that they may all be derived from the complex acoustic reflectance. Specifically, the complex impedance is related to the reflectance via

$$Z(f,x) = r_0\frac{1+\Gamma(f,x)}{1-\Gamma(f,x)}, \tag{5}$$

where the constant r_0 is called the *characteristic acoustic resistance* of the ear canal.

The characteristic resistance is defined as the ratio of pressure to volume velocity for a single wave propagating in the canal. Therefore, it applies to the forward and retrograde waves separately, as follows:

$$r_0 \equiv \frac{P_+(f,x)}{U_+(f,x)} = \frac{P_-(f,x)}{U_-(f,x)} = \frac{\rho c}{A}. \tag{6}$$

Here $\rho \approx 1.2$ kg/m^3 is the density of air, $c \approx 343$ m/s is the speed of sound in air, and A is the ear canal area. The average diameter of the adult ear canal is about 7.5 mm, with an average area of about 44.2×10^{-6} [m^2]. In a real ear, the canal area will vary with distance along the canal, and thus r_0 will also vary. Additionally, the area of the ear canal at the measurement location is typically unknown. Variation due to the use of an incorrect r_0 (Equation 5) has been shown to have a relatively small effect on reflectance and impedance compared to individual variation across ears (Keefe et al., 1992; Voss & Allen, 1994).

Other than for the simplest of cases (e.g., $|\Gamma=0|$), the transformation in Equation 5 can be mathematically challenging. Regardless of the mathematical complexity of the relationship, impedance and reflectance are mathematically equivalent (interchangeable). Since reflectance is more intuitive than other WAI quantities, we focus on reflectance.

Sound Transmission in the Middle Ear

The middle ear converts airborne sound into cochlear fluid-borne sound by the mechanical action of the TM and ossicles. Vibration of the TM drives the ossicles. The ossicles then transmit the signal to the annular ligament of the oval window, which in turn transmits the signal to the fluid-filled cochlea.

Sounds are transmitted efficiently from the canal to the cochlea in the normal middle ear, with little loss in acoustic power (Parent & Allen, 2010; Puria & Allen, 1998). The middle ear may be modeled as a cascade of transmission lines (mathematical models for the flow of acoustic power) that are approximately matched, having relatively little loss and few reflections along the pathway from the ear canal to the inner ear (Allen, 1986; Lüscher & Zwislocki, 1947; Møller, 1983; Puria & Allen, 1991, 1998). If the middle ear becomes unbalanced, abnormal reflections occur, transmission is impaired, and this translates to poorer hearing. Little power is lost for reverse traveling signals (Allen & Fahey, 1992), giving us the opportunity to study both the middle ear and cochlea using acoustic measurements made in the ear canal.

Small vibrations generated in the cochlea by nonlinear motions of the outer hair cells can be measured in the ear canal (Kemp, 1978). These are known as otoacoustic emissions (OAEs), and they are an important tool for studying the function of the inner ear. It is important to know the status of the middle ear when interpreting OAE measurements, since the OAE-evoking acoustic signal must first travel through the middle ear into the cochlea, then the evoked retrograde OAEs must travel back out through the middle ear to the ear canal. An abnormal middle ear will disrupt sound propagation both to and from the cochlea.

The auditory system is exquisitely sensitive. The reference level 0 dB SPL is defined by the threshold of hearing at 1 kHz and corresponds to pressure vibrations of about 2×10^{-5} Pa (for reference, ambient atmospheric pressure is on the order of 10^5 Pa). The middle ear is also remarkably robust, especially compared to the inner ear. The middle ear is not damaged by sounds of extremely high intensity (on the order of 120 dB SPL). The inner ear, in contrast, is subject to substantial damage from sounds of this intensity. The inner ear is protected by at least two efferent systems. The acoustic reflex from the stapedius muscle (Feeney & Keefe, 1999, 2001; Møller, 1983) helps protect the cochlea from intense low-frequency vibrations. The tensor tympani, which connects to the long process of the malleus, may be activated during speaking and chewing (Aron, Floyd, & Bance, 2015; Bance et al., 2013).

The middle ear may be represented as an acoustic transmission line, consisting of compliance (spring), resistance, and mass elements, as shown in Figure 1–2. Each of these elements has an impedance associated with it, which causes sound pressure waves to be reflected in a frequency-dependent manner; the combination of these elements determines the reflectance at the TM. The uniform tubes represent time delay. Figure 1–2 depicts the outer ear (pinna and concha) and ear canal as a series of tubes of varying area. The TM is also a transmission line, with approximately 36 μs of delay (Puria & Allen, 1998). Following are the ossicles, consisting of the masses of the malleus,

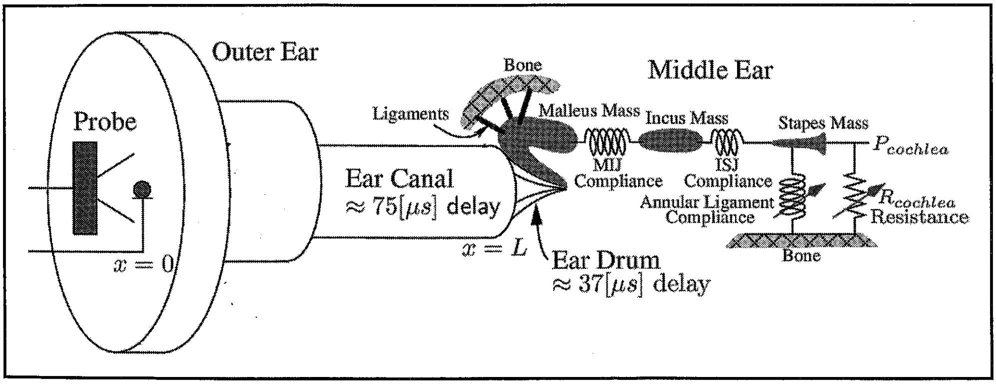

Figure 1–2. A transmission line model of the middle ear is shown, including the outer ear (pinna and concha), the ear canal (represented as a tube), and the TM, which is also a transmission line with approximately 36 μs of delay (Puria & Allen, 1998). Following are the ossicles, consisting of the masses of the malleus, incus, and stapes. Between each of these masses is the joint ligament, which is represented as a spring (MIJ: malleoincudal; ISJ: incudostapedial). The annular ligament holds the stapes footplate in the oval window; this spring is nonlinear since it changes its compliance when force is applied via the stapedius muscle. Finally, the cochlear impedance is represented as a nonlinear resistance. The middle ear, TM, and ear-canal transmission lines are all well matched, such that sound power is delivered efficiently to the ear over a broad range of frequencies (Allen, 1986; Lüscher & Zwislocki, 1947; Møller, 1983; Puria & Allen, 1991, 1998).

incus, and stapes. Between each of these masses is a spring, which represents the compliant ligament between each pair of ossicles. The annular ligament holds the stapes footplate in the oval window; this spring is nonlinear, as its compliance changes when force is applied via the stapedius muscle. Finally, the cochlear impedance is represented as a nonlinear resistance. In general, a nonlinear model element has an impedance (and reflectance) that depends on the stimulus.

Sound Reflections in the Outer and Middle Ear

Impedance discontinuities lead to reflected pressure waves, which contribute to the reflectance measured at the probe. When a sound pressure wave propagates through the middle ear in the model of Figure 1–2, it is modified by changes in impedance that can be complicated functions of frequency.

Compliance (spring) elements, such as the ligaments, have an impedance that is inversely related to frequency; thus they have a high impedance at low frequencies. Mass elements (such as the ossicles) have an impedance that increases linearly with frequency. Finally, ideal resistance elements have impedances that are constant with frequency. A combination of these elements results in a more complicated frequency dependence of pressure reflections.

Reflectance measured at the probe microphone $\Gamma(f,0)$ varies as a function of frequency and depends on how the acoustic impedance of the TM varies with frequency. At frequencies below 1 kHz, the impedance of the TM is due mostly to the compliance (stiffness) of the annular ligament (Allen, 1986; Lynch, Nedzelnitsky, & Peake, 1982; Lynch, Peake, & Rosowski, 1994) and other middle-ear structures. When pressure waves at these low frequencies reach the stapes, almost all of their power is briefly stored as potential energy in the stretched ligament (spring) and then reflected back to the ear canal as a retrograde pressure wave (Allen, Jeng, & Levitt, 2005). At even lower frequencies, below about 800 Hz, only a small fraction of the incident power is absorbed into the middle ear and cochlea (Parent & Allen, 2010; Puria & Allen, 1998).

In a normal ear, in the mid-frequency region between 1 and 4 kHz (or higher), the stiffness- and mass-based impedance effects of the middle ear largely cancel each other. Additionally, the resistance-based impedance in this region has a similar magnitude to that of the stiffness- and mass-based impedances. As a result, much of the incident power that reaches the TM in this frequency region is absorbed into the middle ear and transmitted to the inner ear.

At high frequencies above 6 kHz, the mass-based impedance of the ossicles can dominate the TM impedance (Allen et al., 2005). When a high-frequency pressure wave reaches the TM and mass-based impedance is substantial, most of the power in the incident pressure wave is momentarily stored as kinetic energy, primarily in the ossicles, and then reflected back to the ear canal as a retrograde pressure wave.

The ideal ear canal may be modeled as a rigid-walled (lossless) cylinder of constant area. Under this assumption, there are no reflections along the ear canal except at the TM. In this case, the reflectance at the probe microphone $\Gamma_m(f) = \Gamma(f,0)$ is related to the TM reflectance $\Gamma_{tm}(f) = \Gamma(f,L)$ by a pure delay, expressed as

$$\Gamma_m(f) = \Gamma_{tm}(f)e^{-j2\pi f 2L/c}. \tag{7}$$

There is a round trip delay of $2L/c$ between the measurement point and the TM (Voss & Allen, 1994). Thus, when the ear canal is lossless,

$$|\Gamma_m(f)| = |\Gamma_{tm}(f)|. \tag{8}$$

This has been shown to be a good approximation in adult ears (Voss, Horton, Woodbury, & Sheffield, 2008), though it may be affected by the depth and quality of the probe insertion, and any compliance-related loss associated with the ear canal walls (Abur, Horton, & Voss, 2014).

Given the measured $\Gamma(f,0)$ and a known canal length L, one may estimate $\Gamma_{tm}(f,L)$, and thus the complex TM reflectance. Voss and Allen (1994) used the complex reflectance to estimate the acoustic properties of the TM, by removing pure delay from the reflectance phase. When the area of the canal depends on position (as in a real ear canal), the effective length L is a function of frequency, and removing the effects of the ear canal from the complex reflectance is nontrivial. A number of methods to estimate the ear-canal length and remove ear-canal effects have been proposed (including Keefe, 2007; Lewis & Neely, 2015; Rasetshwane & Neely, 2011; Robinson, Nguyen, & Allen, 2013).

Measurements and Procedures

Applications of Reflectance

Next we discuss some of the main applications of complex reflectance measurements. Many clinical studies to date have considered the reflectance magnitude, in the form of power reflectance or absorbance level, which we describe in the following sections. Additionally, current results indicate that *forward pressure level* (FPL) calibrations allow for accurate individualized calibration of stimuli for assessment of the inner ear (Scheperle, Neely, Kopun, & Gorga, 2008; Withnell, Jeng, Waldvogel, Morgenstein, & Allen, 2009). FPL is derived from pressure reflectance measurements in the ear canal to account for the effect of standing waves in the ear canal, which have large and highly variable frequency effects, dependent on the ear and the probe-insertion depth. Finally, methods to analyze the complex reflectance are presented. These include time- and frequency-domain methods to study the complex reflectance, with a particular focus on how to account for the variable phase effects of the *residual ear canal* between the probe and the TM. For clinical diagnoses, the TM reflectance is the quantity of interest.

Power Reflectance and Absorbance Level

As previously noted, the magnitude reflectance measured at the probe microphone location in the ear canal $|\Gamma_m(f)| = |\Gamma(f,0)|$ may be assumed to be approximately equal to the magnitude reflectance at the TM $|\Gamma_{tm}(f)| = |\Gamma(f,L)|$, while its phase is highly varying across ears. For this reason, clinical studies have focused on the *power reflectance,* $|\Gamma_m(f)|^2$. Because this value ranges between 0 and 1, expressing the ratio of power reflected from the middle ear, it is often expressed as a percentage.

A related quantity, the *power absorbance,* $1 - |\Gamma_m(f)|^2$, is a measure of middle-ear energy transmission, indicating approximately how much power is conveyed to the middle ear and cochlea (Allen et al., 2005; Rosowski et al., 2012). With some middle-ear abnormalities, power can be absorbed by the middle ear and does not reach the cochlea. The power absorbance expressed in decibels with reference to the total absorbance, $10log_{10}(1 - |\Gamma m\,(f)|^2)$, is referred to as the *power absorbance level,* and has a distinctive shape for normal ears, similar to the middle-ear transfer function. This quantity has also been referred to as *transmittance* (Allen et al., 2005).

Figure 1–3 shows example power reflectance and absorbance level measurements from normal ears (Voss & Allen, 1994), an ear simulator (the Brüel & Kjær 4157), and a rigid cylindrical cavity. The cavity has a power reflectance close to 100% across all frequencies, as expected, with some small losses due to viscous and thermal effects of airflow along the cylinder walls (Keefe, 1984). For normal ears, the absorbance level has a distinctive shape, with a rising slope below about 1 kHz, a relatively flat region with very little attenuation (e.g., –3 dB) between about 1 and 4 kHz (Rosowski et al., 2012), and a falling slope at high frequencies. Note that in the mid-frequency range (e.g., 1–4 kHz) large individual variations of the power reflectance (e.g., a 40% range) corresponds to a small range of decibel variation (e.g., 3 dB) of the absorbance level.

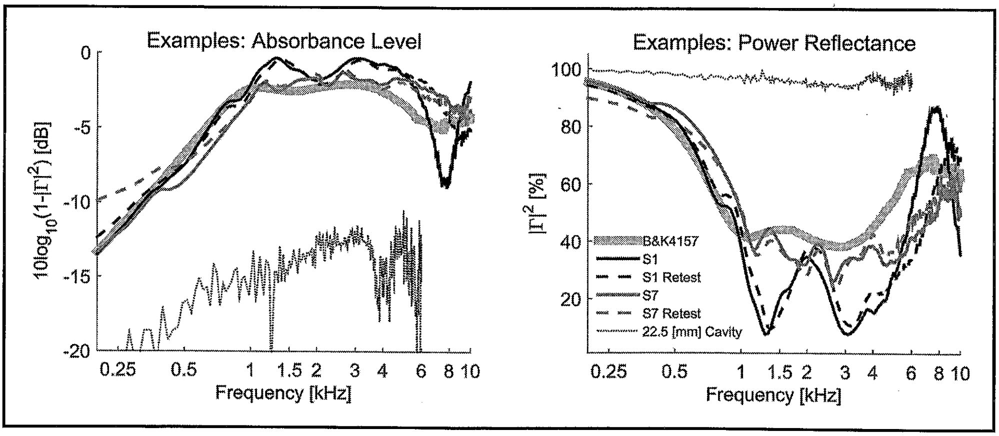

Figure 1–3. Absorbance level (*left*) and power reflectance (*right*) for two normal ears (with retest measurements) from Voss and Allen (1994). Additionally, a measurement of the Brüel & Kjær 4157 ear simulator from that study is presented, where the ear simulator is intended to mimic the response characteristics of the average adult ear. Finally, a measurement of a rigid cylindrical cavity (22.5 mm long) is shown.

Forward Pressure Level (FPL)

Many acoustic assessments of the inner ear, including OAE and hearing thresholds (e.g., the audiogram), rely on the transmission of sound stimuli to the cochlea via the outer and middle ear. Therefore, proper calibration of such stimuli requires an understanding of the magnitude and phase effects introduced by stimulus propagation through the ear canal and middle ear.

Standard practice to account for middle-ear effects is to calibrate stimuli using a middle-ear simulator, often referred to as *reference equivalent threshold sound pressure level* (RETSPL) calibration (ISO, 1997; Souza, Dhar, Neely, & Siegel, 2014). However, ear simulators do not account for the significant variation of middle-ear properties across normal individuals. More important, they do not account for probe-insertion depth, which varies across ears and probe insertions. Probe-insertion depth is of particular importance because acoustic standing waves occur in the ear canal between the measurement probe and the TM (Siegel, 1994).

A standing wave is created when the forward and retrograde pressures in the ear canal are out of phase, nearly canceling each other and creating a deep minimum (e.g., −20 dB) in the total pressure magnitude (Equation 1). The frequency at which this cancellation occurs is dependent upon the round-trip delay from the probe source ($x = 0$) to the TM ($x = L$) and varies considerably across probe insertions and across normal ears. To understand the standing wave effect, consider a rigid cylindrical cavity of uniform area and length L. The retrograde pressure at the microphone is related to the forward pressure by $P_-(f,0) = P_+(f,0)e^{-j2\pi f\tau}$, where the round trip delay is $\tau = 2L/c$. Therefore, the total pressure at the microphone is

$$P_m(f) = P(f,0) = P_+(f,0)(1+e^{-j2\pi f\tau}), \quad (9)$$

which goes to zero for any frequency where $e^{-j2\pi f\tau} = -1$ (which occurs when the quantity $2\pi f\tau$ is equal to π, plus or minus any integer multiple of 2π). In the case of a real-ear measurement, delay from the TM and middle ear will also contribute to the frequency location of the pressure null.

It is now recognized that the *forward pressure level* (FPL) should be used for such stimulus calibrations, to account for standing wave effects on the stimulus magnitude in individual ears (Scheperle et al., 2008; Souza et al., 2014; Withnell et al., 2009). Thus we present an ear-dependent variation of the RETSPL method here, which may be called *reference equivalent forward pressure level* (RETFPL) calibration. At the time of this writing, RETFPL calibration is available to hearing researchers, and will soon be available to clinicians.

The FPL is defined as the forward component of the total pressure wave, $P_+(f,x)$, as previously described. One may determine the forward pressure at the microphone from the total pressure as follows:

$$P_m(f) = P = P_+(f,0) + P_-(f,0) =$$
$$P_+(f,0)\left(1 + \frac{P_-(f,0)}{P_+(f,0)}\right) = P_+(f,0)\,(1 + \Gamma(f,0)).$$

Solving for $P_+(f,0)$ gives

$$P_+(f,0) = \frac{P_m(f)}{1+\Gamma_m(f)}. \quad (10)$$

Given the microphone pressure $P_m(f)$ and the measured complex reflectance $\Gamma_m(f)$, which may both be determined by the reflectance measurement system, the forward pressure at the microphone $P_+(f,0)$ can be estimated (Withnell et al., 2009). To perform the RETFPL stimulus correction for inner ear assessment, the loudspeaker voltage can be varied so that $P_+(f,0)$ is constant at the desired level. The reflectance measured at the probe characterizes individual magnitude and phase properties of both ear canal and middle-ear transmission; without it, the standing wave cannot be precisely removed.

Figure 1–4 shows the magnitude of the normalization factor $(1 + \Gamma_m(f))/2$ in decibels for 10 human ears and 2 ear simulators, computed from the measurements of Voss and Allen (1994). The extra factor of 2 is to compensate for the fact that $\Gamma_m(f)$ goes to 1 as the frequency goes to zero. The use of this factor of 2 (6 dB) is optional, depending on what magnitude correction is desired at low frequencies.

Considering Figure 1–4, at low frequencies, the forward and retrograde pressures are approximately in phase, requiring little correction. Above 3 kHz, the phase of $\Gamma_m(f)$ plays a very important role, as it results in a deep null in the correction factor, due to the ear canal standing wave. The frequency location of the correction factor null

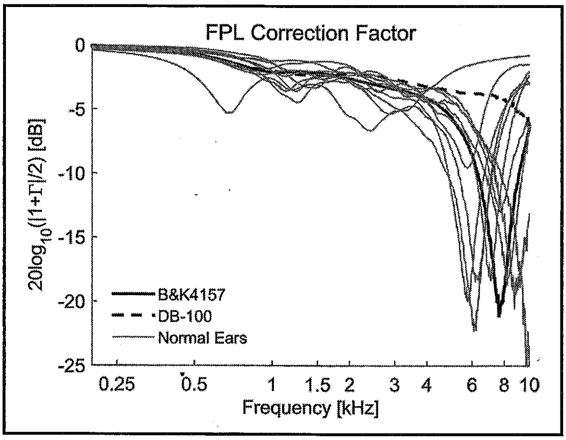

Figure 1–4. Forward pressure level (FPL) normalization factor $|1 + \Gamma_m|/2$, which corrects for the ear canal standing wave, for 10 normal ears and two ear simulators from Voss and Allen (1994). At the frequency of the null, the phase of the complex reflectance is approximately 180 degrees (Withnell et al., 2009). The frequency of the null critically depends on the round-trip delay between the probe tip and the TM (Equation 9). This delay is different for each ear, as it depends on the insertion depth of the probe and the geometry of the ear canal and TM. As the ear-canal delay decreases, the standing wave null shifts upward in frequency.

increases as the distance between the probe and TM decreases. To understand the effect of the TM reflectance on the standing wave null frequency, consider a distance of 1.5 cm between the probe and TM. This gives a round-trip delay of about 87 μs, and thus the standing wave null might be estimated to occur at 5.7 kHz. However, including additional delay from the TM (about 36 μs) and ossicles, the actual null frequency will be lower.

Based on the deep nulls in the FPL correction factor (due to ear canal standing waves) observed in Figure 1–4, it would not be reasonable to normalize the microphone pressure $P_m(f)$ to be constant when delivering stimuli to the cochlea. Such a normalization would boost the level at the standing wave null frequency by as much as 25 dB (in Equation 10, if $P_m(f)$ is held constant there is a peak in the forward pressure corresponding to the minimum of $|1 + \Gamma_m(f)|$). Calibration using an ear simulator will typically not be effective either, even if an artificial ear canal of similar length is included in the measurement. Consider the correction factors for the two ear simulators shown in Figure 1–4. The DB-100 has a very short ear canal, such that the correction factor is nearly constant. In the case of a longer simulated canal, the Brüel & Kjær 4157, it is unlikely that the length of the simulator canal will precisely equal the distance between the probe microphone and TM for the real-ear measurement. Because the correction factor minima are so narrow and deep, this length must be precise to avoid introducing a deep attenuation at the false null frequency, in addition to boosting the FPL much too high at the true null frequency.

Complex Frequency-Domain Reflectance

For middle-ear diagnostics, the quantity of interest is the reflectance of the TM, $\Gamma_{tm}(f) = \Gamma(f,L)$, as a function of frequency. Given $\Gamma_{tm}(f)$, other WAI quantities may be determined at the TM (e.g., Equation 5). As previously discussed, we are not able to measure the reflectance directly at the TM, and the residual ear canal length is unknown (it must be estimated). Therefore, clinical studies to date have considered only the magnitude reflectance (Equation 8), in the form of power reflectance or power absorbance level. However, taking the magnitude of the complex reflectance eliminates phase information from the TM, which may provide additional diagnostic information, when separated from the ear canal phase. Thus, a topic of recent interest is modeling the ear canal contribution to reflectance, with the goal of making the complex TM reflectance available for clinical investigation. Researchers and modelers have approached this with a variety of time- and frequency-domain methods (Keefe, 2007; Lewis & Neely, 2015; Rasetshwane & Neely, 2011; Robinson et al., 2013).

The measured reflectance phase has two additive components: phase from the middle-ear structures (e.g., the TM and ossicles) and ear-canal phase, due to the residual ear canal delay between the probe tip and the TM. The reflectance magnitude has two multiplicative components associated with energy loss and absorbance by the middle ear and ear canal. For the ideal case, where the ear canal is assumed to be a lossless uniform tube, this is described by Equation 7. More generally, we can represent the complex reflectance at the microphone as

$$\Gamma(f) = \Gamma_{tm}(f)\Gamma_{ec}(f). \quad (11)$$

Here $\Gamma_{tm}(f)$ is the complex reflectance at the TM, which is the quantity of interest for middle-ear diagnostics, while the residual ear canal factor, $\Gamma_{ec}(f)$, accounts for round-trip sound propagation in the nonideal ear canal.

Due to nonideal properties of the human ear canal, it is nontrivial to estimate its contribution to the complex reflectance measured at the probe. In a real ear, the phase of $\Gamma_{ec}(f)$ will have a frequency dependence that is related to both the ear canal length and area variation between the probe and TM. Deviation of the magnitude of $\Gamma_{ec}(f)$ from 1 is due to acoustic losses, related to the compliance of the ear canal walls. These losses are assumed to be small in normal adult ears but increase with probe distance from the TM.

Robinson et al. (2013) factored the frequency-domain reflectance such that the ear canal component, $\Gamma_{ec}(f)$, has a magnitude of 1 for all frequencies, assuming the ear canal has no acoustic losses. The method accommodates any lossless delay from an ear canal of varying area (e.g., in Equation 7, the length parameter L may be considered to be a function of frequency). This factorization is unique when performed on a parameterized version of the complex frequency-domain reflectance. Though the ear canal must have at least small losses, this method is consistent with the common assumption used in reflectance analysis (i.e., $|\Gamma(f)| = |\Gamma_{tm}(f)|$), while also providing phase information associated with the TM factor, $\Gamma_{tm}(f)$. In this procedure, any lossless delay associated with the middle ear will also be attributed to the ear canal.

Other studies have considered frequency-domain transmission line models of the ear canal and middle ear (e.g., Figure 1–2), including ear-canal elements of variable area (Lewis & Neely, 2015). It is difficult to verify either of these methods using real-ear measurements, as measurements are highly affected by probe placement within a few mm of the TM, due to the complicated sound field close to the TM.

Time-Domain Reflectance

Time-domain methods have been proposed to analyze the reflectance phase. Due to the variation in distance between the probe and middle-ear structures (e.g., the malleus is closer to the probe than the stapes, along the sound pathway), reflections from these structures may be distinguished based on peaks in the time-domain reflectance (Neely, Stenfelt, & Schairer, 2013).

Using time-domain reflectance, Rasetshwane and Neely (2011) described a method to determine the ear-canal area as a function of distance along the canal. Estimates of the ear-canal area function may be used to model the complex TM reflectance from the reflectance measured at the microphone. For instance, one may approximate the ear canal as a series of concatenated tubes or conic sections of varying area and calculate a transmission line model result, in both the time and frequency domains.

A good time-domain (or frequency-domain) estimate of the TM reflectance contains delay information associated with various middle-ear structures, which may, in addition to the magnitude, be systematically altered in the presence of a middle-ear abnormality.

Wideband Tympanometry

The primary focus of this chapter is reflectance measured in the ear canal at ambient static atmospheric pressure. A clinical alternative to this technique involves pressurizing the ear canal as in tympanometry, to measure the reflectance as a function of frequency *and* ear-canal static pressure. This is typically represented as a three-dimensional magnitude plot (Margolis, Saly, & Keefe, 1999), where the canal pressure is swept from about +200 to −400 daPa. An advantage to this technique is that it combines wideband power absorbance with tympanometry, the standard clinical technology. Disadvantages of this measure include the need to pressurize the ear canal, and the effects pressurization can have on subsequent measurements, due to preconditioning of the TM (Burdiek & Sun, 2014). Wideband tympanometry generates more data to analyze than WAI, which must be interpreted for clinical decision-making, a subject of current research (Keefe, Hunter, Patrick Feeney, & Fitzpatrick, 2015). Clinical efficacy is still being established; for example, Keefe, Sanford, Ellison, Fitzpatrick, and Gorga (2012) did not find an advantage to adding pressurization for detecting CHL in children.

Measurement Technique

System and Calibration

Typically, a reflectance measurement is made by playing a broadband sound stimulus in the ear canal, such as a chirp, and measuring the sound pressure at the probe microphone. The sound source must be calibrated in order to correctly interpret the ear-canal sound pressure response. Most techniques for measuring reflectance use multiple cylindrical cavities of known lengths to calibrate the probe; these approaches differ primarily in terms of the size, length, and the number of calibration cavities used (Keefe et al., 1992; Møller, 1960; Neely & Gorga, 1998). For instance, Mimosa Acoustics' *HearID*® and *OtoStat*® systems (Champaign, IL) uses a four-cavity method described by Allen (1986). In this method, the pressure responses of four cylindrical cavities are compared to their theoretical values, in order to determine the *Thévenin equivalent* source parameters, describing the acoustic behavior of the probe loudspeaker(s).

Reflectance can also be derived using other calibration methods such as the two-microphone method or the standing-wave tube method (Beranek, 1949; Shaw, 1980). The two-microphone method requires a precise calibration of the microphones and precise knowledge of their relative placement; the standing wave tube method requires manual manipulation of the microphone placement with reference to the end of a tube. Many of these early methods have been shown to be relatively inaccurate due to their sensitivity to precise placement of the microphone(s). The four-cavity method is suitable for clinical use because it is less sensitive to precise placement of the microphone.

Any calibration may be verified by measuring a system of known impedance. The simplest method is to measure the reflectance of a syringe or hard-walled lossless cavity. In this case, we expect $|\Gamma_m(f)| \approx 1$ across the frequency range of measurement for an approximately lossless load impedance. Larger errors can occur above about 5 kHz due to more complicated patterns in the ear canal pressure at higher frequencies. Since an ear is not a rigid cavity, perhaps a more easily interpreted test is to measure an artificial ear. However, such a coupler and its expected frequency response may not be readily available, making this approach less practical.

Challenges and Sources of Variability

For investigators working with reflectance in the lab or clinic, it is important to be aware of sources of measurement variability that can be controlled (including calibration, noise, probe ear-tip insertion, and acoustic leaks) and those that cannot (such as normal variation across ears).

Acoustic probe calibration. Acoustic measurements can be very sensitive to probe calibration. In the case of a foam-tipped probe, parameters characterizing the sound source may drift over time with expansion and compression of the foam (ear-tip geometry), and be sensitive to differences across replaceable tips, such as small variations in the tip length. Environmental noise and vibration can affect calibrations and be propagated through to measurements. It is important to always carefully follow the manufacturer's guidelines when calibrating acoustic equipment. Manufacturers are working on systems that can be factory calibrated, removing the need for daily field calibrations.

Noise. Acoustic measurements can be sensitive to environmental noise. In the hospital or clinic, such noise is often unavoidable. If this is the case, some ways to obtain better data include increasing averaging times for data acquisition (if this parameter is under user control) or making repeated measurements. A set of measurements may, in some cases (e.g., the power reflectance), be averaged, or the measurement with lowest noise may be selected during analysis. Often this can be accomplished automatically, via artifact-rejection software included with the measurement device.

In addition to noise in the environment, acoustic noise from vocalizations and movement by the patient may also affect measurements. Data from infants often have the most measurement noise because they are collected in a busy hospital, and the infant cannot be asked to sit still and be quiet during measurement. Noise can also be propagated through the probe's electrical cable if it is draped over the patient. Normative data as well as individual measurements may be degraded by such noise.

Probe insertion. The depth of the probe in the ear canal and the probe seal may both have effects on the measured reflectance, though the effect of probe depth is typically assumed to be small. Power reflectance at low frequencies has been shown to decrease (absorbance level increases) with the probe distance from the eardrum (Lewis & Neely, 2015; Voss et al., 2008; Voss, Stenfelt, Neely, & Rosowski, 2013). This effect is relatively small in adult ears (infant ears have larger ear-canal losses) when the probe is situated in or close to the bony portion of the ear canal, because this region is most similar to a rigid-walled cavity and thus has smaller acoustic losses. Outside of the bony region, the cartilaginous section of the ear canal has much greater wall losses. Therefore, it is best if the probe tip is inserted deeply enough (e.g., 1 cm) to reach the bony part of the ear canal. Probe location in the ear canal may also vary due to different probe-tip types across measurement systems. For instance, "umbrella" tips, which may be used for pressurized wideband tympanometry measurements, are placed against the opening to the ear canal. By construction, measurements made with these tips will not extend as far into the ear canal as an insert ear tip, leading to a much lower ear-canal standing wave frequency and more ear-canal wall losses.

Finally, investigators should have a basic knowledge regarding the effects of acoustic leaks on reflectance, so that a leaky probe insertion may be corrected during the measurement sitting. The measurement device may include methods to detect these leaks, but they may not be effective 100% of the time (e.g., for small leaks). Most leaks occur because the probe tip does not properly seal in the ear canal, or may shift in the ear canal over the course of multiple measurements. Sometimes, using a smaller tip more deeply inserted will decrease issues with leaks.

An example of an acoustic leak is given in Figure 1–5. These measurements are from a normal ear, where the probe insertion drifted over the course of multiple tests made over a period of a few minutes (reflectance data from

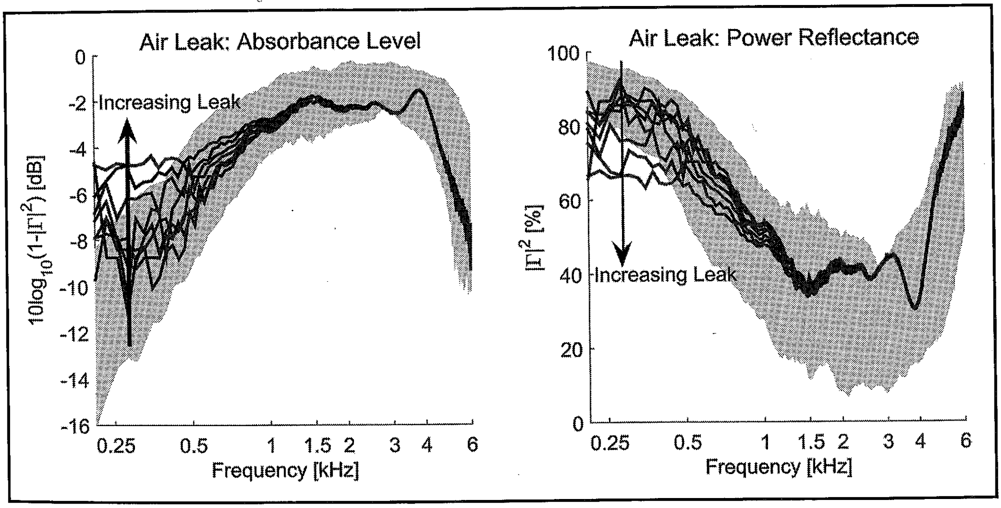

Figure 1–5. Example of the effects of a leak in probe insertion on the absorbance level and power reflectance in a normal ear, plotted against the 10th to 90th percentile (*gray region*) for normal ears from Rosowski et al. (2012). As the size of the acoustic leak increases (e.g., the probe insertion slowly loosens), there is an increase in the low-frequency absorbance level and a decrease in the low-frequency power reflectance. The effect propagates upward in frequency as the leak increases.

Thompson, 2013). As the leak in the probe seal increases in size, the low-frequency absorbance level increases, and the corresponding power reflectance curve decreases (Groon, Rasetshwane, Kopun, Gorga, & Neely, 2015). This increased absorbance is not from the middle ear absorbing the acoustic power but from the power dissipating through the leak around the probe tip. For most middle-ear conditions, unless the TM is abnormally compliant or perforated, the absorbance level should be relatively low (power reflectance close to one, or 100%) below 1 kHz. Groon et al. (2015) recommend that when the frequency range of interest extends as low as 0.1 kHz, low-frequency absorbance should be ≤0.20 and low-frequency admittance phase ≥61 degrees; for frequency ranges as low as 0.2 kHz, low-frequency absorbance should be ≤0.29 and low-frequency admittance phase ≥44 degrees.

Normal variation. Normal middle ears are known to have a fairly wide range of variation in power reflectance and absorbance level (Rosowski et al., 2012). The largest source of *intra*subject variability (i.e., test-retest within the same ear) is probe placement in the ear canal (Voss et al., 2013), though this variability is small compared to variability across a population of normal ears (Rosowski et al., 2012). Middle-ear pressure within a normal range may also cause intrasubject variations (Shaver & Sun, 2013).

When the probe is properly placed in the ear canal, *inter*subject variation (i.e., across ears) is due to differences in middle-ear physiology. Voss et al. (2008) found, based on manipulations in cadaveric ears, that variations in the volume of the middle-ear space produced larger variability in power reflectance measurements than variation in probe insertions. The middle-ear cavity can affect the power reflectance and absorbance level over a broad frequency range and may play a role in variability across normal subjects. Variability of the absorbance level (see Figure 1–3) between 1 to 4 kHz is small (e.g., ±3 dB) in the normal ear, and it is likely related to the acoustics of the TM and ossicles, in addition to middle-ear space (Rosowski et al., 2012; Stepp & Voss, 2005).

Clinical Applications

The middle ear is a complex mechanism with many components. It follows that there are many possible disorders of the middle ear. Some disorders include fluid or infection in the middle-ear space, ossification of the bony structures, discontinuities of the ossicular chain, perforation of the eardrum, and various abnormalities of the membranes, ligaments, and supporting structures. Wideband reflectance offers a novel approach to describe and diagnose middle-ear dysfunction. Although complex pressure reflectance offers a more complete picture, to date, clinical researchers have focused on power reflectance and power absorbance. These quantities contain similar information, and they are both widely used. We provide both power reflectance and absorbance where possible to help the reader navigate the literature. Arguably, the power absorbance or absorbance level is the preferable format, due to its close relationship with the middle-ear response when plotted in decibels.

There are many approaches to establish diagnostic criteria. Statistical methods are used to establish the normal range and to identify criteria for what is abnormal. Mathematical models are used to simulate and describe the underlining physics of normal variability and abnormal changes in the middle ear (Parent & Allen, 2010; Voss, Merchant, & Horton, 2012). Combining these methods allows us to derive useful diagnostic criteria. In the discussions below, different studies demonstrate different approaches to establish such criteria. In some cases, the diagnostic criteria are built into the measurement software, and in other cases, the clinician or researcher may need to derive the result from exported data. The clinical utility of WAI is still undergoing intensive research, and clinical applications are rapidly advancing.

Quantities Used

A single WAI measurement produces a wealth of information. The measurement technique provides frequency resolution on the order of 20 Hz over at least a 0.2- to 6-kHz range. Typically, this is too much information to work with statistically for the purpose of clinical decision making. Thus, clinical researchers use various tactics to reduce the number of variables and extract meaningful quantities to assist diagnostic decision making. Current approaches include:

1. *Looking for patterns.* Small-*N* and case studies are used to get an idea of the general pattern of normal and abnormal results, particularly when characterizing relatively unstudied pathologies. This can help focus attention onto specific frequency ranges in larger studies and improve detection based on physical modeling.
2. *Band averaging.* Band-averaging power absorbance level and power reflectance across frequency can describe frequency-dependent behavior using a smaller set of parameters (e.g., the reflectance area index (RAI) defined in Hunter, Feeney, Lapsley Miller, Jeng, & Bohning, 2010). One-third, one-half, and whole-octave bands are frequently used and fit nicely with other audiologic tests, while still capturing the shape of the power absorbance curve. It is not typically useful to take the average across the entire curve, because frequency-dependent behavior is obscured.
3. *Comparison to norms.* Comparing abnormal results to a norm can be done qualitatively and quantitatively. For instance, the Absorbance Level Difference (ALD), defined by Rosowski et al. (2012), is the absorbance level relative to a normal ear average over a specific frequency range. They used the ALD to quantify notches seen in abnormal absorbance curves.
4. *Parameterization.* The WAI response may be characterized with a small number of parameters, such as a three-line approximation to the absorbance level curve (Rosowski et al., 2012).
5. *Multivariate approaches.* Methods such as discriminant function analysis and multiple regression can help the researcher narrow down variables that provide the most unique information. These

approaches can also allow information to be combined across test types and across measurements with different units, providing a powerful basis for clinical decision making. Multiple parameters are combined to produce one number, which is then used for making a decision.

6. *Physical models.* Parameters are extracted using a physics-based model for the middle ear. For example, Robinson, Thompson, and Allen (in press) and Lewis and Neely (2015) extract parameters from simple transmission line models of the middle ear to characterize the condition of the middle ear.

Norms

For clinical use, we must first establish the normal range of immittance quantities, broken down by key demographics such as sex, ear, age, and ethnicity. With information about how normal ears behave, ears with middle-ear dysfunction may be identified. A norm is a statistically-defined range for a given quantity, derived from highly screened normal ears from a specific demographic group. The type and degree of screening for defining "normal" varies across studies, but includes audiologic history and audiologic tests such as auditory brainstem response (ABR), OAE, tympanometry, surgical discovery, and pneumatic otoscopy. Different screening tests and criteria can lead to differences between norms across studies. Norms can be expressed in various ways, including percentile ranges and means with standard deviations.

As summarized by Shahnaz, Feeney, and Schairer (2013), the demographic that has the greatest effect on the normal middle ear is age, followed to a much lesser extent by ethnicity. Sex and ear differences are much smaller and typically not statistically significant. Figure 1–6 shows norms for four key age groups, replotted from their original studies in consistent units (median and 10th–90th percentile range) to aid comparison. Here we focus on the age demographic and describe some key normative studies, with additional discussion about secondary demographic differences. Across many studies, the general consensus is that larger-*N* normative studies are needed using highly screened normal ears. Many existing norms are based on small *N*, especially by the time they are divided into various demographic groupings. Notable gaps include older children and teenagers, and the elderly. There has been a focus on infants and young children due to their high prevalence of middle-ear disorders.

Norms may obscure individual patterns across frequency and can also obscure the noise and variation seen in WAI measurements taken in real-world clinical settings where noise levels may be difficult to control. This is why we have overlaid the norms in Figure 1–6 with five randomly selected examples of normal ears from the same data sets.

Newborn and Infant Norms

The human ear undergoes significant maturation in the first 12 months of life (as summarized in Kei et al., 2013). They found that the largest changes in absorbance and reflectance norms occur between birth and 6 months of age,

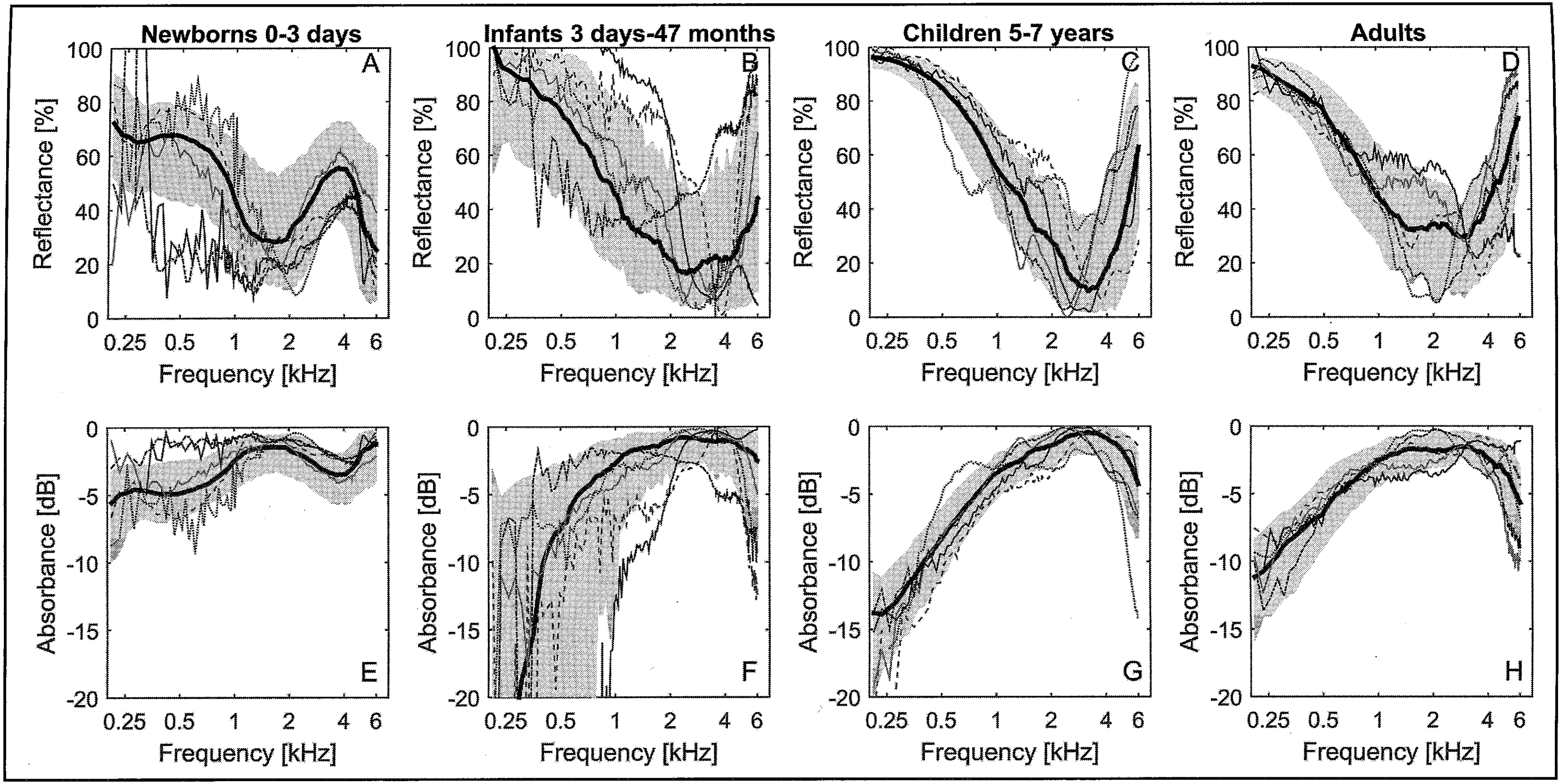

Figure 1–6. Example norms for four age groups, including 10th to 90th percentiles (*gray region*), median (*solid black line*), and five randomly chosen examples (*dotted lines*) for each group. The top row shows power reflectance (%) and the bottom row shows the same data plotted as power absorbance level (dB re 100% Absorbance). These norms were recalculated and replotted from data published elsewhere: Newborn norms are from Hunter et al. (2008); infant norms are from Hunter, Tubaugh, et al. (2008); child norms are from Beers et al. (2010); adult norms are from Rosowski et al. (2012).

indicating that this is the period of most rapid change in the outer and middle ear. Aithal, Kei, and Driscoll (2014b) showed that developmental changes in the outer and middle ear over the first 6 months of life cause a decrease in absorbance for low to mid-frequencies and an increase in absorbance at higher frequencies (>2.5 kHz). The absorbance at low frequencies is dominated by compliance characteristics of the ear canal and middle ear. With maturation, ossification of the inner two thirds of the ear canal causes the ear canal to be less compliant, corresponding to a lower middle-ear absorbance at low frequencies (Kei et al., 2013). Additionally, changes in ossicle bone density with age, along with the loss of mesenchyme and other middle-ear fluids, lead to decreased mass in the middle-ear system (Aithal et al., 2014b; Kei et al., 2013). This leads to an increase in the middle-ear absorbance at high frequencies, as the lower mass causes less sound to be reflected. This rapid maturation implies norms are potentially needed for many age ranges.

Norms from newborn babies in the first hours to first days of life are of particular interest due to large-scale newborn hearing screening programs that test babies soon after birth. Newborn absorbance level norms typically show unreliable results below 1 kHz due to environmental noise and ear-tip leaks, and high absorbance at 1 to 2 kHz (higher than in older ears), decreasing at 3 to 4 kHz, and rising again at 6 kHz (which is also not seen in norms for older ears). Figure 1–6 shows norms from the Hunter et al. (2010) study, recalculated to show the entire frequency range and to show absorbance. Similar norms were also shown by others (Aithal, Kei, Driscoll, & Khan, 2013; Sanford et al., 2009; Shahnaz, 2008), with differences mainly due to screening criteria for "normal," demographics, and equipment.

Aithal et al. (2013) pointed out that using just distortion product otoacoustic emission (DPOAE) pass/refer results (as the earlier studies did) was not sufficient as a gold standard for normal ears, because ears with strong DPOAEs can overcome middle-ear dysfunction; however, they found similar norms with a smaller, more highly screened group. For newborns, there is ambiguity in defining the "normal" condition, because it is natural for healthy newborns to have some fluid in their external and middle ears, which affects middle-ear measurements. Whether this is an issue or not depends on the purpose. If the aim is to understand the normal infant middle ear, it is important. But if the purpose is to assess infant inner-ear status (as in universal newborn hearing screening [UNHS] programs), any temporary middle-ear condition that affects sound propagation is of concern regardless of whether it is "normal" or not.

Sex and ear differences are typically not observed or are not clinically significant; however, differences in ethnicity were found by Aithal, Kei, and Driscoll (2014a) where Australian Aboriginal infants had lower wideband absorbance than Australian Caucasian infants. This may be of clinical importance due to the high otitis media with effusion (OME) prevalence among Aboriginal children.

For slightly older infants, Hunter, Tubaugh, Jackson, and Propes (2008) produced norms for infants 3 days to 47 months old. They did not find significant age or sex effects, although their age bands had only around 10 ears per band.

Children

The study with the largest number of normative subjects for young children is Beers, Shahnaz, Westerberg, and Kozak (2010). They tested wideband reflectance in 78 children (144 ears) ages 5 to 7 with normal middle-ear function for comparison to those with OME. They compared Caucasian and Chinese children's ears, and found significant differences at 2 and 6 kHz; 2 kHz falls within an important frequency range for detecting conditions that increase middle-ear stiffness, like OME, so this may be of clinical significance. It remains to be studied if body size is a better predictor of variation in absorbance and reflectance than ethnicity.

Adults

Rosowski et al. (2012) established norms on a medium-sized group of highly screened otologically normal adults (29 adults/58 ears, up to age 64). They found small sex and ear differences, and their overall average power reflectance curve was similar to previous studies of power absorbance and reflectance (see Figure 1–6). Of interest is the parameterization of the absorbance curve. In log-log coordinates, the curve can be modeled with three straight lines (Allen et al., 2005; Rosowski et al., 2012). Below 1 kHz, average absorbance increases by about 15 dB per decade. Above 4 kHz, absorbance decreases by 23 dB per decade. Between 1 and 4 kHz, absorbance is essentially constant at around −2.5 dB. Extracting the key features of the absorbance curves and deriving parametric values to characterize them could aid in clinical decision making. For instance, changes in slopes, frequency of intercepts or large deviation from a straight line may be indicative of abnormal middle-ear performance.

Middle-Ear Dysfunction

A number of small-*N* and case studies have suggested where power absorbance and reflectance might be useful for detecting middle-ear dysfunction, and they provide a descriptive patterns of the conditions relative to norms (e.g., Allen et al., 2005; Feeney, Grant, & Marryott, 2003; Sanford & Brockett, 2014). These studies suggest areas where larger studies should look. However, we need larger-*N* studies to understand how middle-ear power absorbance behaves statistically over a population, for each pathology and demographic. These results must then be combined and reduced to specific criteria for decision making. As well as larger-N studies for specific pathologies, broad studies across a range of confusable pathologies are needed to establish the bases for differential diagnoses. Fortunately, a number of such studies are in progress, and we will summarize some here.

Wideband Reflectance in Universal Newborn Hearing Screening (UNHS) Programs

The goal of UNHS programs is to detect babies who have sensorineural hearing loss so they can benefit from early intervention (e.g., cochlear implants). UNHS programs provide a pass or refer result from either OAE or ABR tests. These screening tests are not diagnostic but are used to determine referrals for more extensive diagnostic follow-ups.

It has long been best practice in UNHS programs to rescreen babies who get a refer result to reduce false positives for diagnostic referrals. This rescreening is usually done after a delay, because testing within 24 hours of birth is much more likely to produce a refer result than testing after 24 hours (and preferably 36 hours). The majority of these false-positive referrals are from transient middle-ear dysfunction from the birth process (e.g., amniotic fluid, mesenchyme, and meconium in the middle-ear space), which clears within the first few days of life. Figure 1–7A–D shows examples of five newborn ears from Hunter et al. (2010) that did not pass DPOAE testing. In all cases, their middle-ear reflectance was higher than normal (absorbance was lower than normal), indicating their DPOAE refer result was possibly a false positive. Hunter et al. (2010) showed why rescreening OAEs after a delay was often successful—middle-ear absorbance tends to increase over time, presumably as the middle ear clears, allowing more sound to propagate into the inner ear and back.

This transient middle-ear dysfunction is not reliably picked up with tympanometry in newborns. Hunter et al. (2010) and Sanford et al. (2009) both showed that wideband power absorbance and reflectance are superior to tympanometry in predicting which ears show low OAE levels due to middle-ear dysfunction. Adding absorbance to OAE screening can potentially identify those babies most in need of diagnostic follow-up, help determine the best time for repeat screening, and reduce false-alarm referrals.

WAI can be used to interpret OAE results in older infants, children, and adults similarly to newborns. In these older age groups, however, low absorbance is cause for follow-up for middle-ear dysfunction like otitis media.

WAI in UNHS

When using WAI with OAEs in testing newborns, a pass/refer result can be assigned to each test, giving four possible outcomes. Each outcome is illustrated in Figure 1–8 with real examples from real babies in the Hunter et al. (2010) study. This study showed that the reflectance or absorbance around 2 kHz was the best predictor of a DPOAE test pass (DPOAE levels at 3 or 4 out of 4 frequencies are normal) or refer (DPOAEs at 2 or more out of 4 frequencies are abnormally low). Absorbance below 1 kHz is not plotted because in babies, this region is often noisy (Hunter et al., 2010).

Also plotted are two normative regions in gray. For the absorbance plot, the gray norm represents an

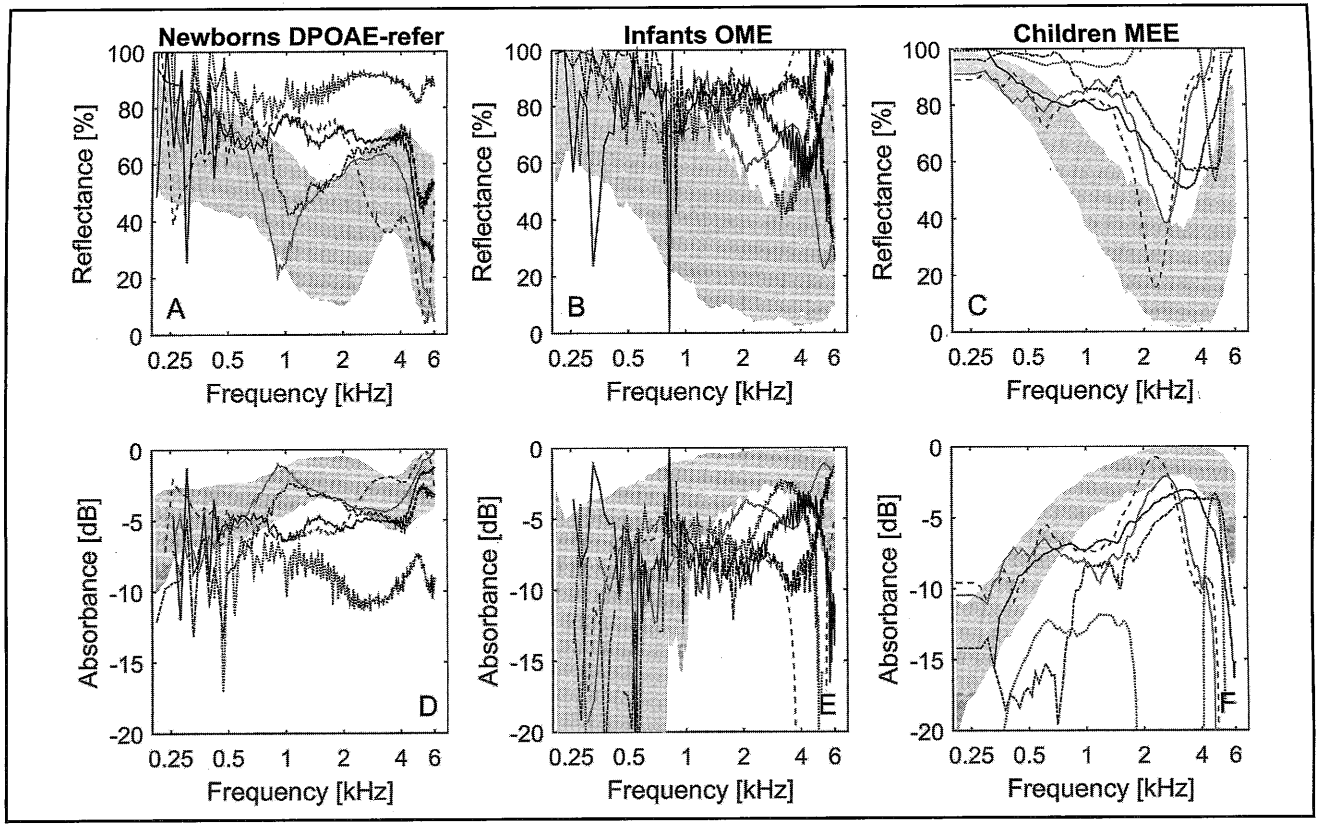

Figure 1–7. Five examples of middle-ear dysfunction in children, overlaying norms from Figure 1–6 (10th–90th percentiles, *gray region*). The top row shows power reflectance (%) and the bottom row shows the same data plotted as power absorbance level (dB re 100% Absorbance). **A–D:** newborns who received a DPOAE refer result (Hunter et al., 2010). **B–E:** infants diagnosed with otitis media with effusion (OME) (Hunter, Bagger-Sjoback, & Lundberg, 2008). **C–F:** children diagnosed with middle-ear effusion (MEE) (Beers et al., 2010).

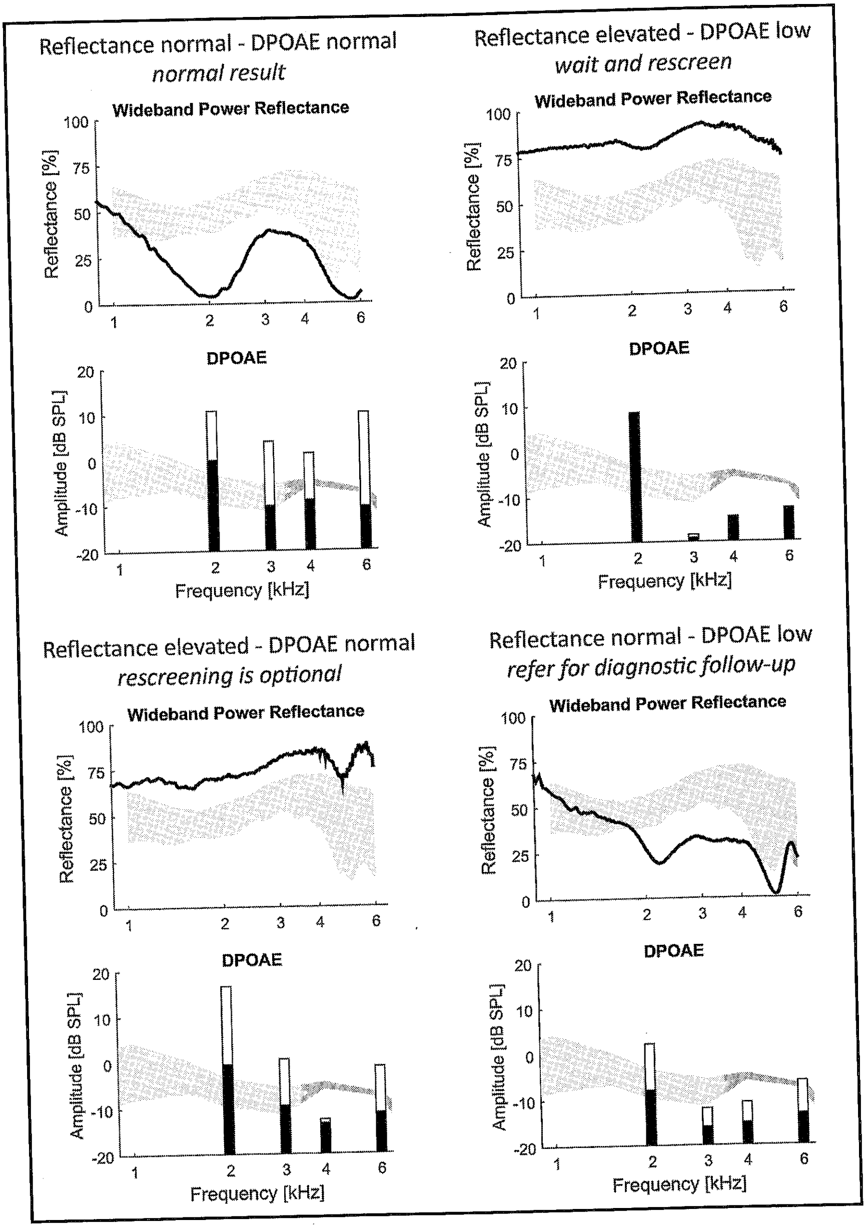

Figure 1–8. Four outcomes are possible when using wideband absorbance at 2 kHz with DPOAEs in a newborn hearing screening program. The absorbance plots show power absorbance (*black line*, dB re 100% absorbed) and the normative ambiguous region (*gray region*, dB re 100% absorbed). The DPOAE plots show DPOAE amplitude (*white bar*, dB SPL), the noise floor (*black bar*, dB SPL), and the Boystown 90% ambiguous region (*gray region*, dB SPL), where DPOAEs below the region are considered refer results. For the screening protocol used in this study, if 3 or 4 out of 4 DPOAE frequencies get a pass result, the overall result is a pass (*left plots*). If 2 or more out of 4 DPOAE frequencies get a refer result or are noisy, the overall result is a DPOAE refer (*right plots*). Referrals can be reduced by considering absorbance. If absorbance is low, the DPOAE refer is probably due to middle-ear dysfunction (*top right*). However, the possibility of underlying sensorineural hearing loss cannot be excluded, so repeat screening is needed. If absorbance is normal, the DPOAE refer needs diagnostic follow-up for possible sensorineural hearing loss (*bottom right*). Sometimes absorbance will be low but the DPOAEs are so strong, they are able to overcome the reduction in middle-ear transmission (*bottom left*).

ambiguous region. Absorbance *above* this region (especially at 2 kHz) was associated with DPOAE pass results. Absorbance *below* this region (especially at 2 kHz) was associated with DPOAE refer results. The ambiguous region describes where the pass and refer regions overlap (defined by the 10th and 90th percentiles). Similarly, for the DPOAE plot, the gray norm also represents an ambiguous region (from the Boystown norms Gorga et al., 1997). DPOAEs *above* this region are associated with normal hearing. DPOAEs *below* this region are associated with abnormal hearing.

How could WAI be used in UNHS programs? Specific guidelines are still in development. Potentially, WAI may be used to enable better timing for rescreening and follow-ups. The examples in Figure 1–8 suggest the following courses of action:

1. Normal absorbance—normal DPOAEs across all frequencies (top left). Screening passed and no rescreening or follow-up is needed.
2. Low absorbance—low DPOAEs: possible middle-ear fluid (top right). Wait a few hours and rescreen to see if absorbance is higher and DPOAEs pass. Refer for diagnostic follow-up if DPOAEs do not pass on rescreening. Chances are absorbance will increase as the middle ear clears and the true DPOAE status will be more clearly revealed. Low absorbance and low DPOAEs are commonly seen in newborn hearing screening programs and cause undue worry for parents and an increased workload due to unnecessary diagnostic follow-ups. With WAI + DPOAEs, testers can immediately see if there is middle-ear dysfunction and can reassure parents that this is common and not of concern.
3. Normal absorbance—low DPOAEs (bottom right). This ear is a priority for diagnostic follow-up because there may be permanent sensorineural hearing loss. Rescreening is optional because the usual reason for DPOAE false alarms—low middle-ear absorbance from transient middle-ear dysfunction—has been eliminated. Any rescreening can occur immediately because the WAI results show the middle ear is not impeding sound propagation into the inner ear.
4. Low absorbance—normal DPOAEs (bottom left). The DPOAEs are strong enough to overcome what is possibly a probe blockage or transient middle-ear dysfunction. In this situation, the tester should check for probe or ear canal blockage, or a collapsed ear canal, and then retest WAI. Since DPOAEs passed, rescreening is optional because an outer or middle-ear condition is not typically a reason for referral.

Identifying Conductive Hearing Loss (CHL) in Infants and Children

Identifying CHL in young infants can be difficult with tympanometry, and there is no standard interpretation. Prieve, Vander Werff, Preston, and Georgantas (2013) evaluated tympanometry variations along with wideband reflectance and showed the latter was just as effective as tympanometry in identifying CHL in infants less than 6 months old (3–26 weeks) who had been referred in an infant hearing screening program. The babies received both air and bone conducted ABR tests along with tympanometry and WAI (43 ears had normal hearing and 17 ears had CHL, determined from the air- and bone-conducted ABR thresholds). Prieve et al. (2013) found that wideband reflectance between 800 and 3000 Hz was higher in CHL ears compared with normal ears (absorbance was lower than normal). Prieve et al. found that a criterion for power reflectance greater than 69% (power absorbance less than 31%) in the one-third octave band around 1,600 Hz produced the highest likelihood ratio for CHL, compared to other frequency bands and compared to various quantities derived from multifrequency tympanometry (at 226, 678, and 1,000 Hz). These results indicate the frequency range most sensitive to CHL in infants and show that power absorbance is a suitable replacement for tympanometry in this age group. A larger study will be needed to determine statistically meaningful criteria, and investigate if further age-specific criteria are needed.

In children, otitis media is the most common reason for CHL. Otitis media can produce middle-ear effusion (MEE) and negative middle-ear pressure (NMEP), which both tend to stiffen the middle ear and thereby decrease the amount of power absorbed for low- to mid- frequencies below 2 to 4 kHz, compared to normal ears (Robinson et al., in press). The extent to which this stiffness affects the mid-frequency range (i.e., between 1 and 4 kHz) may depend on the severity of the condition. For MEE, there can be an additional decrease in absorbance at mid- to high frequencies due to the mass of the fluid. These decreases in absorbance can be used to identify MEE. Hunter, Tubaugh, et al. (2008) point out that tympanometry is unreliable in very young infants since it can produce normal results in the presence of MEE. They found that absorbance was decreased between 1 and 3 kHz in ears with suspected MEE. Figure 1–7 shows five examples of OME from this study, compared to the norms. Below 1 kHz, the data are noisy due to the difficulty in getting quiet measurements on this population. Above 1 kHz there is a decrease in absorbance, consistent with stiffening of the middle ear from OME. Noise rejection has been improved in clinical equipment since this study was conducted.

Beers et al. (2010) tested 78 children (144 ears) with normal middle ears and 64 children with abnormal middle ears (21 ears with suspected MEE, 21 ears with confirmed MEE, and 54 ears with abnormal NMEP). The children were aged 5 to 7 years. They found that reflectance in the frequency region around 1.25 kHz best separated the normal ears from those with MEE. Using the 90th percentile from the normal group as a criterion, all ears with MEE had higher reflectance (lower absorbance) than the criterion at 1.25 kHz (hit rate 100%), for a false-alarm rate of 10%. Figure 1–7 shows five examples of children with MEE from this study, overlaid on the norms, showing greatly increased reflectance and reduced absorbance. They also showed that WAI was much more sensitive than 226-Hz tympanometry in detecting MEE. On average, ears with abnormal NMEP had higher reflectance (lower absorbance) than normal but not as high as those ears with MEE. Ellison et al. (2012) also investigated MEE in children and found similar results to Beers et al. (2010), with decreased absorbance between 1.5 and 3 kHz in" ears with surgically verified MEE.

Middle-Ear Pathology in Adults

In infants and children, the main interest is hearing screening and otitis media. In adults, a range of other middle-ear conditions are common. Figure 1–9 shows five examples of four common conditions, compared to adult norms.

Ossicular Disarticulation/ Discontinuity

The disruption or near-disruption in an ossicular joint causes a significant peak in the absorbance, typically below 1 kHz (Panels A, E in Figure 1–9). This peak is likely due to a resonance of the ossicle mass and TM stiffness. Although the absorbance is elevated in this narrow frequency band, this power is *not* transmitted to the inner ear but is dissipated in the resonant joint.

Otosclerosis

Stapes fixation (due to otosclerosis) typically reduces the absorbance at low frequencies below 1 to 2 kHz but may be within normal limits, or even lower than normal. Figure 1–9 (Panels B, F) shows five examples from Shahnaz et al. (2009) that illustrate these three outcomes. In the typical otosclerotic ear, Shahnaz et al. (2009) found increased reflectance (decreased absorbance) below 1 kHz, compared to a normal group. This is due to increased stiffness of the middle-ear at the stapes, which increases the total stiffness of the middle ear measured at the eardrum. They found that WAI was a better predictor than tympanometry, with an 82% hit rate and 17% false alarm rate, and identified 500 Hz as a good band for detecting otosclerotic ears. They also identified a subgroup of ears where reflectance was lower than normal (absorbance was higher than normal) below 1 kHz. A combination of WAI and tympanometry was able to detect all otosclerotic ears for the price of a higher false-alarm rate.

Tympanic Membrane (TM) Perforations or Pressure Equalization (PE) Tubes

TM perforations or pressure-equalization (PE) tubes tend to produce power absorbance curves that are highly variable across frequency, with high absorbance at low frequencies and a large equivalent ear canal volume, typically much greater than 3 cc. It can be difficult to differentiate a noisy test with an acoustic leak from a TM perforation; however, repeated testing should reveal a stable pattern for PE tubes or a perforation. This variation across ears is apparent in Figure 1–9 (Panels C, G) where five ears with varying degrees of perforation show low reflectance at low frequencies but not much consistency.

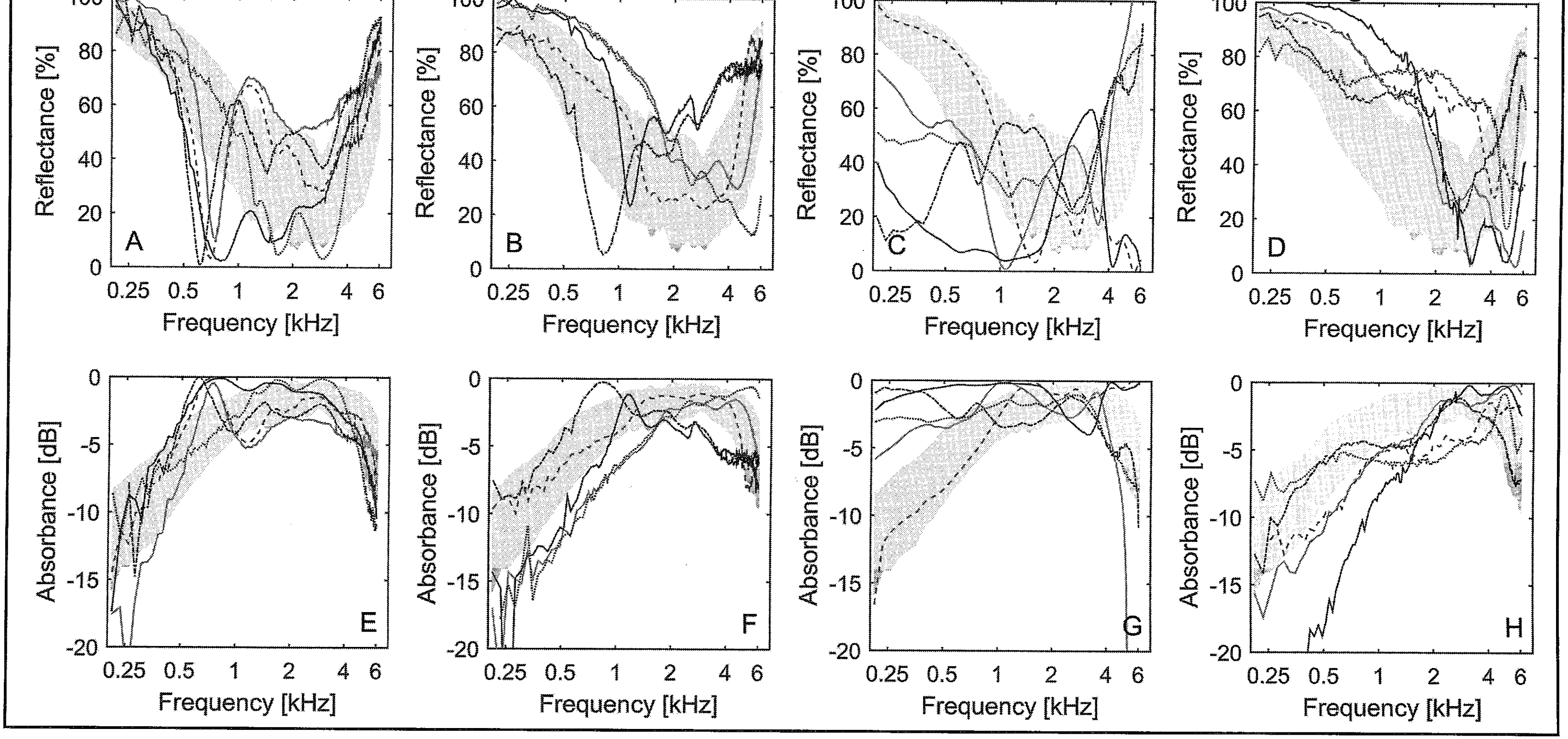

Figure 1–9. Examples of common middle-ear pathologies in adults, overlaid on Rosowski et al. (2012) norms, plotted as power reflectance (%) and absorbance level (dB re 100% absorbance). **A, E:** ossicular discontinuity showing the characteristic notch below 1 kHz (three randomly selected examples from Nakajima). **B, F:** otosclerosis (five randomly selected from Nakajima et al. (2012). **C, G:** TM perforation (five examples from unpublished data). **D, H:** NMEP ranging from approximately −384 to −65 daPa (five examples from Robinson et al. [in press]).

These data come from an unpublished study (Feeney, Hunter, Jeng, & Lapsley Miller).

As in the case of ossicular discontinuity, although the absorbance is elevated due to a TM perforation, this power is *not* transmitted to the inner ear but is dissipated in the middle-ear cavity (Voss, Rosowski, Merchant, & Peake, 2001). The resonant frequency at which we see dissipation is related to the middle-ear cavity size *and* the size of the hole (these two parameters can be modeled as a *Helmholtz resonator* (Voss et al., 2001), which is what you get when you blow over the top of an empty bottle).

Larger perforations are detectable with otoscopy, but smaller perforations may be hard to visualize. In a cadaveric ear, smaller perforations were more apparent in the WAI response than larger ones (Nakajima, Rosowski, Shahnaz, & Voss, 2013; Voss et al., 2012), presenting as very low reflectance around 1 kHz.

Negative Middle-Ear Pressure (NMEP) and Eustachian Tube Dysfunction (ETD) in Adults

Abnormal NMEP can occur from eustachian tube dysfunction (ETD) and it typically increases the stiffness of the middle ear. On average, NMEP causes a decrease in the power absorbance level for low- to mid-frequencies, below 2 to 4 kHz, with large intersubject variation in NMEP effects. Across ears and NMEP levels, the absorbance is most sensitive to NMEP near 1 kHz (Robinson et al., in press; Shaver & Sun, 2013). Middle-ear pressure cannot be directly estimated from WAI as is the case with tympanometry (i.e., from tympanic peak pressure readings). The degree of absorbance change is associated with degree of NMEP, but this is most noticeable when considering changes in a subject or in large group averages—it is not easy to detect the degree of NMEP from single measurements unless it is severe. It is important to evaluate the presence of NMEP as it can affect other measurements, especially OAEs (even with NMEP within a clinically normal range). Figure 1–9 (Panels D, H) illustrates five examples of ears with NMEP where normal-hearing experimental subjects induced NMEP using a Toynbee maneuver (closing mouth and pinching nostrils, then swallowing) (Robinson et al., in press). Tympanometry indicated NMEP ranging from approximately −384 to −65 daPa.

Differential Diagnosis of Conductive Hearing Loss (CHL) in Adults

It should now be apparent that although some middle-ear pathologies are easy to detect from normal, in some cases they can be difficult to distinguish because the effects on WAI are similar, despite having different causes. For instance, otosclerosis and NMEP can both cause an increased stiffness as seen at the eardrum even though that stiffness is generated in different ways. From what we know so far about WAI, it is not always possible to *differentially* diagnose every middle-ear disorder. However, in conjunction with other tests (such as air-bone gap [ABG]), different disorders can be teased out, including some conditions where it had been previously difficult, expensive, or not possible.

An excellent example is CHL with an intact TM and aerated middle ear, which can be associated with three conditions: ossicular fixation (usually from otosclerosis), ossicular discontinuity, and superior semicircular canal dehiscence (SCD) (Nakajima et al., 2012). These three conditions are challenging to differentially diagnose in the clinic and may require surgery or expensive tests to fully investigate. Nakajima et al. (2012) showed how to use WAI and an ABG audiogram to aid in differential diagnosis in an office setting.

Ossicular discontinuity and otosclerosis are described above. SCD is often referred to as a "third window" lesion of the inner ear (i.e., in addition to the oval and round windows of the cochlea), caused by a space where there is bone loss (Merchant & Rosowski, 2008). This affects the reflection of sound from the cochlea. SCD ears tend to show an abnormal peak in the absorbance around 1 kHz (Nakajima et al., 2012), which is smaller, wider, and higher in frequency than typically seen with ossicular discontinuity.

The absorbance level difference (ALD) used by Nakajima et al. (2012) for differential diagnosis was calculated by subtracting the band-averaged absorbance level (in decibels) over 0.6 to 1 kHz from the mean normative absorbance level (averaged over the same range) from the companion study of normal ears by Rosowski et al. (2012) The ABG they used for the differential diagnosis was defined as the average gap between 1 and 4 kHz (this separates out the SCD cases where the ABG is most apparent at frequencies less than 1 kHz). This test is applicable to patients presenting with CHL, defined as >10 dB ABG on pure-tone audiometry (averaged over 500, 1,000, 2,000 Hz or 250, 500, 1,000 Hz), and with an intact TM and aerated middle ear. Nakajima et al. (2012) found:

- Ears with ALD (0.6–1 kHz) <1 dB and ABG (1–4 kHz) >10 dB were associated with stapes fixation.
- Ears with ABG (1–4 kHz) ≤10 dB were associated with SCD.
- Ears with ALD (0.6–1 kHz) ≥1 dB and ABG (1–4 kHz) >20 dB were associated with ossicular discontinuity.

In this study, sensitivity and specificity were good (stapes fixation: 86%/100%, ossicular discontinuity: 83%/96%, and SCD: 100%/95% for sensitivity and specificity, respectively). These numbers are based on $N = 31$ ears. These results suggest further study is warranted to refine the differential diagnostic potential of WAI and audiometry.

The Past and the Future of Wideband Acoustic Immittance (WAI)

There is a long history of effort in developing measurement methods and instrumentation for middle-ear evaluation. It took 30 years for tympanometry to become part of the standard clinical battery for hearing evaluation. In light of the high newborn hearing screening false-positive rate and the advantages of digital technology, the development of WAI as a clinical tool for middle-ear evaluation was launched with the support of National Institute on Deafness & Other Communication Disorders (NIDCD) funding in the mid-1990s. The first commercial research WAI measurement system was made available in 2003, and the first clinical instrument with Food and Drug Administration (FDA) clearance was available in 2006. As each WAI measurement provides an abundance of information compared to tympanometry, WAI opened wide a window into middle-ear assessment.

There are several important clinical applications that WAI can support. Some clinical applications of WAI go beyond the scope of this chapter. For instance, WAI may be used to assess middle-ear acoustic reflex. It has been demonstrated that the middle-ear reflex threshold obtained using WAI is lower than that obtained using standard tympanometry (Schairer, Feeney, & Sanford, 2013). Another area of application is detection of traumatic brain injury; WAI measurements may be compared before and after a traumatic incident, or used to monitor recovery (Voss et al., 2010).

In this chapter, we discussed two important clinical applications of WAI: RETFPL stimulus calibration, and noninvasive middle-ear assessment and diagnosis. Stimulus level adjustment based on FPL provides individualized in-the-ear calibrations, such that acoustic stimuli may be delivered to the inner ear at the intended sound pressure level. This will enable more accurate and repeatable hearing assessments for all audiologic tests, and more accurate performance for hearing instruments (e.g., hearing aids). Additionally, if an RETFPL calibration is used for OAE measurement, WAI is acquired without any extra effort and may be used to inform the diagnosis.

WAI may currently be combined with other audiologic tests to provide quick differential diagnosis for hearing screening and for advanced hearing assessment. Many studies have systematic changes in WAI between normal and pathological middle ears, and made recommendations for detection of certain pathologies. This is especially valuable for pathologies that require middle-ear surgery, where the ability to more accurately identify the pathology and its degree before surgery will improve the surgery preparation and patient care (in some cases, unnecessary exploratory surgery can be avoided). WAI can also be used for monitoring middle-ear status pre- and post-surgery.

Though many studies to date considered only the power absorbance or reflectance, due to unknown phase contributions of the residual ear canal between the probe and TM, many strategies for estimating the complex TM reflectance have been proposed based on physical models of the ear canal. Analysis of the TM reflectance phase is expected to improve differential diagnosis in the future. In addition to advances in modeling middle-ear function, larger normative populations are needed to improve clinical viability of WAI for differential diagnosis of middle-ear pathology.

The clinical applications of WAI are many, and the technology is ready.

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APPENDIX 1–A
Reflective Terminology

ole defines the terminology re- › acoustic reflectance. The um- term *wideband acoustic immit-* WAI) refers to all flavors of ance, admittance, and reflec- Aost impedance concepts may erstood in terms of reflectance, is conceptually equivalent to ance but a more intuitive con- on. When transforming an im- e (or admittance) to reflec- one must first normalize it by canal characteristic resistance *A(x)*, where r_0 is the density of c is the speed of sound, and A is the cross-sectional area of the ear canal. Working with the normalized impedance (admittance) simplifies the interpretation of WAI data. Furthermore, it reduces the variability across subjects, since the area is best estimated when the data is taken, based on the size of the probe tip used for the measurement. These measures may be expressed either in the time domain (e.g., $p(t,x)$), which is real, or in the frequency domain (e.g., $P(f,x)$), which is complex. The term "mho" is "ohm" spelled backwards.

P Cha
p
v
p
e
co refl

Physical Characteristic	Term	Function Representation	Unit
pressure	sound pressure	$p(t,x)$ or $P(f,x)$ [Pa]	pascal
	sound pressure level (SPL)	$20 \log_{10} \left(\frac{P(f,x)}{P_{ref}}\right)$ [dB SPL] $P_{ref} = 20 * 10^{-6}$ [Pa]	decibel
velocity	particle velocity (SVL)	$v(t,x)$ or $V(f,x)$ [m/s]	meter per second
	volume velocity	$U(f,x) = A(x)V(f,x)$ [m^3/s]	cubic meter per second
power	sound intensity	$\mathbb{i}(t,x) = p(t,x)\,v(t,x)$ [W/m^2]	watt per square meter
	sound intensity level (SIL)	$10 \log_{10} \left(\frac{\mathbb{i}(t,x)}{\mathbb{i}_{ref}}\right)$ [dB SIL] $\mathbb{i}_{ref} = 10^{-12}$ [Pa]	decibel
	sound power	$p(t,x) = \mathbb{i}(t,x)A(x)$ [W] or $P(f,x)$ [W]	watt
energy	sound energy	$\varepsilon(t,x) = \int_{-\infty}^{t} \mathbb{i}(\tau,x)A(x)d\tau$ [J]	joule
complex reflectance $\Gamma(f)$	reflectance magnitude	$\lvert\Gamma(f,x)\rvert$	ratio (dimensionless)
	reflectance phase	$\angle\Gamma(f,x)$ [rad]	radian
	reflectance group delay (phase slope)	$\tau_\Gamma(f,x) = \frac{-1}{2\pi}\frac{d}{df}\angle\Gamma(f,x)$ [s]	second
	power reflectance	$\lvert\Gamma(f,x)\rvert^2$	ratio (dimensionless)
	power absorbance	$1 - \lvert\Gamma(f,x)\rvert^2$	ratio (dimensionless)
	power absorbance level	$10\log_{10}(1 - \lvert\Gamma(f,x)\rvert^2)$ [dB]	decibel

continues

ppendix 1–A. *continued*

Physical Characteristic	Term	Function Representation	Unit
complex admittance $Y(f)$	admittance magnitude	$\lvert Y(f,x)\rvert$ [S or $\mho$]	siemens or mho
	admittance phase	$\angle Y(f,x)$ [rad]	radian
	conductance (real part)	$G(f,x) = Re\{Y(f,x)\}$ [S or $\mho$]	siemens or mho
	susceptance (imaginary part)	$B(f,x) = Im\{Y(f,x)\}$ [S or $\mho$]	siemens or mho
complex impedance $Z(f)$	impedance magnitude	$\lvert Z(f,x)\rvert$ [Ω]	ohm
	impedance phase	$\angle Z(f,x)$ [rad]	radian
	resistance (real part)	$R(f,x) = Re\{Z(f,x)\}$ [Ω]	ohm
	reactance (imaginary part)	$X(f,x) = Im\{Z(f,x)\}$ [Ω]	ohm