Wideband acoustic immittance: Concepts and clinical utility

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Abstract

Measurements of middle ear (ME) acoustic power flow and related measures (e.g., reflectance, absorbance, normalized impedance, acoustic resistance, and acoustic reactance) have several important advantages over other measures of acoustic impedance, such as the clinical “gold standard” measure, tympanometry. In addition to its wide bandwidth, ease of use, the absorbance curve, on a dB scale, has a shape that is similar to the ME transfer function, and is easily codified by three audiologically relevant parameters. The wideband acoustic impedance measures (resistance and reactance) can provide additional important information about ME status. Alone, or together with other audiological tests such as otoacoustic emission tests and audiometry, acoustical tympanic membrane (TM) measurements can help identify many abnormal conditions, including degrees of otitis media, TM perforations, otosclerosis and ossicular disarticulation. In addition, the problem of ear canal standing waves may be solved by the introduction of the Forward Pressure level (FPL) calibration, Reference Equivalent Forward Pressure Level (RETFPL) as an alternative to the present “gold standard” Reference Equivalent Sound Pressure Level (RETSPL). Reflectance/absorbance also provides new ways of measuring the middle-ear reflex (MER) that may be more sensitive, thus allowing lower stimulus levels. Normative data from several studies, are summarized.

Abbreviations: DPOAE = distortion product otoacoustic emission; HL = hearing level; ME = middle ear; MER = middle ear reflex; OME = otitis media with effusion; RETSPL = Reference equivalent sound pressure level; RETFPL = Reference equivalent forward pressure level; SPL = sound pressure level; FPL = Forward pressure level; Tympanic membrane = TM.

Key words: acoustic impedance, acoustic power reflectance, characteristic impedance, conductive disorders, middle ear pathologies, otitis media, otoacoustic emissions, otosclerosis, transmittance.

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1 Introduction

When an acoustic pressure wave travels down the ear canal toward the TM, the frequency-dependent power $P(f)$ [W],\(^1\) which is defined as the flow of energy over time and space, is continuous until it reaches an impedance discontinuity, such as a change in the area of the ear canal, or the tympanic membrane (TM). Such variations in the wave propagation result in frequency dependent reflections. The ratio of the reflected to absorbed power is called the power reflectance. The ratio of the reflected to absorbed pressure (or velocity) is called the complex pressure reflectance. Historically reflectance is denoted by the upper case Greek letter “gamma” $\Gamma(f)$.\(^2\)

1.1 Absorbed and reflected waves

The key behind the reflectance measure is to decompose the acoustic wave into absorbed and reflected components. The notation is simplified if we relabel the absorbed and reflected waves in terms of their direction of travel. Thus we define pressures $P_+(f) = P_a(f)$ and $P_-(f) = P_r(f)$ where the “±” indicates the direction of travel of the wave. This results in the pressure and velocity relations

$$ P(f) = P_+(f) + P_-(f) \quad (1) $$

and

$$ U(f) = U_+(f) - U_-(f). \quad (2) $$

The pressure (like voltage) is a scalar, while the velocity (like current), is a vector, with direction. This explains the change in sign of Eq. 2.

At each point $x$ along the canal, the ratios of the pressures $P_\pm$ and velocities $U_\pm$ are equal to the acoustic resistance, which due to conservation of energy must always be positive and real. Thus

$$ \frac{\rho c}{A(x)} = \frac{P_+(f)}{U_+(f)} = \frac{P_-(f)}{U_-(f)}, \quad (3) $$

where $\rho c = 407$ [Rayls] and $A(x)$ is the canal area at location $x$ along the ear canal.

If we reorganize the above equation we obtain the definition of the complex reflectance

$$ \Gamma(f) = \frac{P_-(f)}{P_+(f)} = \frac{U_-(f)}{U_+(f)}, \quad (4) $$

in terms of the ratio of reflected over absorbed pressure (or velocity). This relationship is complex, thus $\Gamma$ may be written either as the sum of a real and imaginary part, or in terms of its magnitude and phase$^3$ (Eq. A.4).

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\(^1\)For clarity all SI units will be displayed in square brackets.

\(^2\)This article assumes the reader knows mathematics at the level of high school algebra. In keeping with this assumption, we will use mathematical notation, which frequently includes Greek letters. Using mathematical notation is simpler because it precisely specifies the variables via a unique symbol.

\(^3\)Given the nature of the topic it is necessary to use complex numbers to represent pressure and velocity, and their ratio, which defines the impedance at a given frequency. For example, the algebraic sentence $\Gamma(f) = |\Gamma(f)| e^{j\phi}$ says that the complex reflectance $\Gamma(f)$, as a function of frequency $f$, is equal to the magnitude of $\Gamma$ at each frequency, times the exponent of the phase $\phi$, or angle. Here $e^{j\phi} = \cos(\phi) + j \sin(\phi)$ is called Euler’s formula ($j = \sqrt{-1}$).
The beauty of complex reflectance over the complex impedance is that the magnitude reflectance $|\Gamma(f)|$ describes the relative absorbed power as a function of frequency, while the phase codifies the latency of the absorbed power (e.g., depth of the reflected wave). Furthermore complex reflectance has certain properties that render it more useful than impedance in forming a diagnosis of a ME pathology.

Thus the complex reflectance has an intuitive interpretation, whereas most people find complex impedance difficult to master. Within the clinical audiology community it is well known that impedance is a difficult concept. Part of the problem is the obtuse terminology, as outlined in Table B1. But terminology is not the entire story. Even those who say they understand complex impedance find it difficult to explain to lay persons. One obvious reason for this is that the transformation from complex reflectance to complex impedance is complicated [Voss and Allen, 1994]. Other than for the simplest of cases, this transformation is mathematically challenging.\(^4\)

It should be noted however that, regardless of the complexity of the relationship, complex impedance and reflectance are mathematically equivalent. Since reflectance is easily understood, and impedance is not, we shall focus on the complex reflectance in this report.

Estimating $\Gamma(f)$ requires a special calibration of the earphone, known in the Engineering literature as a Thévenin calibration. While quite important, this measure is beyond the scope of the present discussion.\(^5\)

Voss and Allen [1994] were the first to use the complex reflectance to characterize the acoustic properties of the TM, by approximately removing the effect of the residual ear canal. When the area of the canal is constant the TM reflectance $\Gamma_{tm}(f)$ is related to the measured microphone reflectance $\Gamma_m(f)$ by the relation

$$\Gamma_m(f) = \Gamma_{tm} e^{-j2\pi f 2L/c}$$ (5)

where $L$ is the distance from microphone to TM and $c$ is the speed of sound. Simply stated, this equation says that there is a round trip delay of $2L/c$ between the measurement point and the TM. Thus if one can estimate $L$ they then can compute $\Gamma_{tm}(f)$, which is mathematically equivalent to the TM impedance. But there is one further complication: When the area of the canal depends on position, the effective length $L$ depends on frequency. This complicates this relationship, for both practice and theory.

Thus if one could estimate $L$, then one may estimate $\Gamma_{tm}(f)$ from the measured $\Gamma_m(f)$. An accurate procedure for doing this that includes the case of variable canal area was reported by Robinson and Allen [2013]. For most clinical applications $\Gamma_{tm}(f)$ is the starting point for the analysis of ME pathologies.

The remainder of this chapter is focused on clinical applications of complex reflectance and the effect of ME pathologies.

1.2 Clinical applications of the complex reflectance

Next we discuss some of the main applications of the complex reflectance method. The first is the Forward pressure level which is a method for calibrating the signal in the ear canal so that the pressure delivered to the cochlea is approximately constant, thus

\[^4\text{This transformation is known as a Möbius, or bilinear, transformation.}\]

\[^5\text{http://en.wikipedia.org/wiki/Thevenin’s_theorem}\]
removing troubling ear canal standing waves. A second application is a method to remove the effect of the residual ear canal, defined as the section of canal between the microphone and the TM. Clinically, what we really need to know is the reflectance at the TM. This is distorted by the round trip delay of the residual ear canal. Finally, given the complex reflectance at the TM, it is possible, in theory at least, to diagnose most, if not all, middle ear pathologies.

What the research clinician would most like to know is the transmission properties of the ME. Specifically is the ME clinically normal? If not, in what way is it abnormal? There is a host of possible abnormalities. The most common are: 1) TM abnormalities, 2) Ossicular disruption and ossification, 3) Fluid in the middle ear, 4) Eustachian tube dysfunction. There is building evidence that the TM complex reflectance can be used to both identify and then quantify (diagnose) most, if not all of these pathologies. We shall present the evidence in support of this hypothesis.

**Forward pressure level and its applications:** Real ear measurement systems use a microphone in the ear canal to account for variation in the subject’s ME. This methods is intended to replace traditional but less reliable coupler calibrations, which do not account for these natural variations. In principle this seems like a logical thing to do. However there are several serious problems that need to be addressed. The most obvious of these is the effect of standing waves in the canal [Siegel, 1994]. While the canal standing wave problem has been recognized for a long time, there have been years of debate as to how to best deal with it. There was considerable uncertainty in the beginning, as to the best quantity to normalize, with the intensity being one obvious candidate (Gorga and Neely, 1998; Souza, 2014; Keefe et al.).

The nature of these standing waves is shown in Fig. 1. Based on the deep nulls seen in the figure, clearly would not be reasonable to normalize the microphone pressure $P_m(f)$ to be a constant. Such a normalization would boost the level at the standing wave null by as much as 25 [dB]. The frequency of this boost would depend on the microphone placement. However this is precisely what is done in real ear measurements when the microphone level is re-normalized to be constant. Real-ear calibrations have the unintended effect of putting a large peak at the standing wave frequency.

![Figure 1: Forward pressure level normalization factor which corrects the ear canal standing wave. At the null frequency $f_0$, the phase of the complex reflectance (i.e., the angle of $\Gamma(f_0)$) is 180 degrees [Withnell, 2010, 2009b]. This explains the source of the standing wave nulls, and their unpredictable nature, since the frequency critically depend on the round trip delay from the source back to the microphone (Eq. 5). This delay is different for each ear, as it dependent on the insertion depth of the probe and the geometry of the canal and TM.](image-url)
community. It is now recognized that the forward pressure level (FPL) should be made constant [Souza, Dhar, Neely, Siegel 2014 (9 methods; JASA)]. However, other than some research systems, FPL has not made it into clinical equipment at the time of this writing.

This calibration procedure is a variant on the so-called RETSPL method, where the voltage on the transducer is varied to maintain the pressure at the TM constant in an artificial ear. Here we coin the term RETFPL (reference equivalent forward pressure level). The procedure is: Given the microphone pressure \( P_m(f) \) and its corresponding complex reflectance \( \Gamma_m(f) \), compute the forward pressure \( P_+(f) \). Vary the receiver voltage so that \( P_+(f) \) is constant at the desired level.

One may determine \( P_+(f) \) as follows:

\[
P_+(f) = P_+(f) + P_-(f) = P_+(f) \left( 1 + \frac{P_-}{P_+} \right) = P_+(f)(1 + \Gamma_m(f)).
\]

Solving for \( P_+ \) gives

\[
P_+(f) = \frac{P_m(f)}{1 + \Gamma_m(f)}.
\]

Since the microphone pressure and the complex reflectance are both known, one may easily solve for the absorbed (i.e., forward) pressure by the above relation. While some alternative methods have been tried, it seems unlikely that, without knowledge of the complex reflectance \( \Gamma(f) \), the precision needed to estimate the standing wave frequency will be adequate.

Fig. 6 shows the normalization factor for 10 human ears, taken from [Voss and Allen, 1994]. As the frequency is increased above 3 [kHz], the phase plays a very important role, as it can result in ear canal standing waves, as seen by the canal microphone. This phase relation is negligible at low frequencies, where it is small. At higher frequencies, the phase, due to the round trip delay from Tagus to TM (2.5 cm) of \( \approx 150 \) [\( \mu \)s], gives a standing wave null at \( \approx 3.4 \) [kHz]. If one includes the additional delay in the TM (\( \approx 45 \) [\( \mu \)s]) and ossicles, the frequency null can be even lower. Examples of these nulls, as shown in Fig. 1, have been computed from the complex reflectance using the relation

\[
FPL = 20 \log_{10} \left| \frac{1 + \Gamma(f)}{2} \right|.
\]

The extra factor of 2 is to compensate for the fact that \( \Gamma(f) \) goes to 1 as the frequency goes to zero. Thus the sum of the two terms is 2, which is removed by this factor. The use of this extra factor is optional depending on the interpretation of what is desired at low frequencies. This factor is known in the literature as the missing 6 dB.
**In summary:** When a earphone excites sound in the ear canal, a forward traveling wave is launched into the canal. This wave proceeds down the canal at the speed of sound. Once it reaches the TM (or any other change in impedance) it is reflected by an amount given by the TM reflection coefficient $\Gamma_{tm}(f)$, in a frequency-dependent manner. This results in a backward propagating wave $P_-(f) = \Gamma_{tm}(f)P_+(f)$. All of the acoustic physics of the TM is captured by $\Gamma_m(f)$. When the microphone measurement location is not at the TM, there is a phase factor given by the round-trip delay between the TM and the microphone location which is the source of the standing wave.

**TM Acoustic admittance:** From the clinical diagnostic point of view, the most important quantity is the acoustic admittance of the TM $Y_{tm}(f)$. This is what tympanometry does not, and cannot accurately estimate. While the low frequency canal admittance magnitude may be measured (this is what 226 [Hz] tympanometry provides), $Y_{tm}(f)$ is not measured at relevant (i.e., speech) frequencies. Furthermore, the complex TM admittance is not accurately estimated from the canal admittance, even at 226 [Hz], because the clinical procedure does not accurately remove the effect of the residual ear canal [Shanks et al, 1981; Rabowitz, 1981; Shanks et al., 1988].

There are several challenges here. Clinically speaking, we would like to estimate the wideband complex admittance at the TM given pressure measurements at some unknown location in the canal. Note we do not know either 1) the length of the ear canal, nor 2) the pressure at the TM. One might wonder if this is even a realistic goal.

What research clinicians would most like to know is the *wideband complex reflectance* at the TM, $\Gamma_{tm}(f) = P_-(f)/P_+(f)$. As previously stated, given $\Gamma_m(f)$ the complex TM admittance or impedance are easily determined. The proposal then is to Thevenin calibrate the ear canal probe. Then given the ear canal pressure $P_m(f)$ one may compute the complex reflectance $\Gamma_m(f)$. Using the procedure of Robinson and Allen (2013), the phase factor corresponding to the residual ear canal may be uniquely determined and removed, leaving the desired $\Gamma_{tm}(f)$. This then puts us at the starting point for the ME diagnosis, as reviewed next.

### 1.3 Overview of Middle Ear Impedance and Reflectance

The middle ear is remarkable for its functionality in converting air-born sound into fluid-born sound, via the mechanical action of the TM and ossicles. Air-born sound that reaches the TM is converted to acoustic vibration of the TM. This vibration drives the ossicles that transmit the acoustic signal to the annular ligament in the oval window which, in turn, convey the signal to the fluid filled cochlea.

**A middle ear model** The middle ear may be represented as a simple circuit, as shown in Fig. 2. At low frequencies (i.e, 200 [Hz]), the delay may be ignored, and this model reduces to a parallel combination of the compliance of the ear canal, the compliance of the ossicular joints and the compliance of the annular ligament. Around 0.7-1 [kHz] the impedance of the cochlea, which is well approximated by a constant resistance, is approximately equal to the impedance of the annular ligament, bringing the cochlear load into the picture. At higher frequencies, the delays of the several transmission lines may not be ignored and the full model must be considered.
Figure 2: This shows a transmission line model of the middle ear including the outer ear (pinna and concha), the ear canal, represented as a tube, the tympanic membrane (ear drum), which is also a transmission line with approximately 37 [µs] of delay. Following this are the ossicles consisting of the mass of the malleus, incus and stapes, shown here as electrical inductors. Between each of these mass’s is the ligament which is represented as a spring, or in the electrical analog, a capacitor. 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 resistance. This is also nonlinear in that it generates acoustic otoacoustic emissions, which are minuscule nonlinear retrograde signals (i.e., OAEs), close to the threshold of hearing for the normal ear. In this figure quote delay across and along canal. Add speaker source, for clarity, and to help raise the concept of the Thevenin source pressure and impedance, due to loading of the speaker by the ear canal. The speaker source impedance/admittance must be close to real given its very small size and internal structure [Kim].

The rapid growth in the use of otoacoustic emissions (OAEs) for hearing screening and related diagnostic evaluations, has focused attention on the need for improved methods of ME assessment [2]. The status of the ME is of key importance for the measurement of otoacoustic emissions, since the external signal must first travel into the cochlea, followed by the evoked retrograde otoacoustic emissions. Both are subject to attenuation from an abnormal ME.

In the normal ME sounds are thus transmitted efficiently from the canal to the cochlea. Relatively little acoustic power is lost in the normal middle ear (<3 [dB]) [Allen, 1986]. Research has shown that the middle ear is a cascade of transmission lines that are approximately matched [Zwislocki, 1947; Moller 1965; Allen, 1986; Puria and Allen, 1991, 1998]. This is expected, since loss would translate to poorer hearing. When this cascade of transmission becomes unbalanced, reflections occur, and transmission is impaired. For this reason the middle ear transfer of energy is very sensitive to any unbalanced impedance elements within the ME structure.

The auditory system is extremely sensitive. The threshold of hearing at 1 [kHz] is approximately 0 sound pressure level [dB-SPL]. This intensity corresponds to an average motion of the stapes of ~8 [pm] (i.e., 1/3 the radius of the Hydrogen atom (25 [pm])).

Due to the reciprocal nature of the middle ear, little power is lost for retrograde signals; i.e., signals traveling in the reverse direction, from the oval window to the TM, have similar loss characteristic as the forward traveling waves [Allen and Fahey, 1992]. It follows that minute vibrations generated in the cochlea by nonlinear motions of the outer hair cells, can be measured in the ear canal. The measurement of these signals, known as otoacoustic emissions (OAEs), are an important tool for studying the mechanism of hearing. This has led to the development of powerful new techniques for the objective assessment of hearing, such as cost-effective methods of world-wide hearing screening.

These signals were discovered by David Kemp and Duck on Kim, as first reported in the early 1970s.
programs.

The ME is also remarkably robust. Sounds of extremely high intensity (on the order of 120 dB SPL) do not damage it. The inner ear, in contrast, is subject to substantial damage from sounds of this intensity (the cilia of the outer hair cells are most easily damaged, which results in sensorineural hearing loss). The ME provides an effective protective mechanism in the form of two middle ear efferent systems. The best understood is the acoustic reflex via the stapedus muscle, which helps protect the cochlea from intense low-frequency vibrations. This mechanism protects the inner ear from intense air-borne sound, but only to a limited extent; the acoustic reflex is too slow to protect the inner ear from intense sounds of short duration [Add references on combat blast levels observed, which can destroy the TM, i.e., 160 dB]. A second efferent system is the tensor tympani, that connects to the long process of the malleus. This system may be used during speaking and chewing.

1.4 Impedance Discontinuities cause Reflected Acoustic Power

The mechanical load on the TM is from the ossicles; thus, when one says the impedance of the middle ear, what one really means is the ear-canal impedance at the microphone location, which is a delayed version of the drum impedance that includes the impedance load of the ossicles and cochlea (Fig. 2). Power reflectance 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 stiffness of the annular ligament [Lynch, 1981; Allen, 1986; 13]. 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 and then reflected back to the ear canal as a retrograde pressure wave. At even lower frequencies (<0.8 [kHz]), only a small fraction of the incident power is absorbed into the ME [14]. The impedance of the TM in this frequency region essentially consists of a stiffness-based reactance.

In a normal ear, in the mid-frequency region between 1 and 5 [kHz], the stiffness- and mass-based reactances of the ME interact in a complex way and largely cancel each other. The ME is a complex structure (see Fig. 2) that consists of several stiffness components and several mass components that all work in harmony such that the cancellation of reactance occurs over a wide frequency range (1 to 5 [kHz]). As a result, most of the incident power that reaches the TM in this region is absorbed into the ME and transmitted to the inner ear. In the low- and high-frequency regions the TM resistance is typically small compared with its reactance, whereas in the mid-frequency region the resistance is larger than the combination of the stiffness- and mass-based reactances.

The stiffness based reactance is inversely proportional to frequency (i.e., it is halved with each doubling of frequency), while the resistance is constant across frequency. For frequencies at the lower end of the auditory range (e.g., 0.1 [kHz]), the reactance is more than 10 times the resistance, but in the region of 1 [kHz], the reactance and resistance are of comparable magnitude. At frequencies above 6 [kHz], the mass-based reactance of the ossicles becomes increasingly important and, because it is linearly proportional to frequency, it can eventually dominate the TM impedance. This is complicated by the fact that the center of mass of each ossicle barely moves. Thus the relevant mass is not the ossicle mass, rather it is the rotational inertia of each ossicle.

To further complicate matters, our experimental knowledge of TM impedance is rel-
atively poor at frequencies above 6 [kHz]; thus, the frequency at which mass-based reactance becomes the dominant component of TM impedance typically is unknown. When a high-frequency pressure wave reaches the TM and mass-based reactance 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. At high frequencies, much of the published data shows an impedance that approaches that of a mass as the frequency is increased. Mass-based reactance is linearly proportional to frequency (i.e., it doubles with each doubling of frequency), while cochlear resistance varies only slightly with frequency [15]. Between 1 and 5 [kHz] the resistance and reactance are of comparable magnitude, whereas at very high frequencies the mass reactance can be several times greater than the cochlear resistance.

2 Methods

2.1 General properties of reflectance and admittance

Definitions of acoustic variables: The acoustic intensity of sound in the ear canal is $I(t) = p(t)u(t)$ [W/m²], where $p(t)$ is the pressure in Pascals [Pa] as functions of time $t$. The acoustic power $P$ is the acoustic intensity $I$ times the cross-sectional area $A$ [m²] of the ear canal, namely $P(t) = A(x)I(t)$ [W]. In general the area varies along the length of the canal $x$ [m]. The acoustic particle velocity [m/s] times the canal area $A(x)$ is called the volume velocity [m³/s].

The intensity and power may be defined either in the time or frequency domains. It is important to be aware of which domain (time or frequency) is being discussed, as these are very different ways of looking at $I$ and $P$. Ideally we should have different notation for intensity and power in the time and frequency domains. Here we shall work almost exclusively in the frequency domain, where all variables shall be defined in terms of a pure tone of magnitude and phase [radians].

The ratio of the pressure over the volume velocity is the acoustic impedance, which for a simple plane wave in the ear canal is $\rho c = 407/A(x)$ [Acoustic Ohms], where $c = 343$ [m/s] is the speed of sound and $\rho = 1.2$ [kgm/m³] is the density of air. For reference, 1 Pascal [Pa] is 94 [dB-SPL], which is typical of yelling. Typical speech is close to 65 [dB-SPL] 1 [m] from the mouth. The average threshold of hearing defines 0 [dB-SPL] which is 20 [µPa], i.e., 20 × 10⁻⁶ [Pa]. If voltage is something you feel you understand, the think of the pressure as “acoustic volts.” Details of these relations are further discussed in Appendix A.

When the ear canal area is constant, the propagation delay is constant and the phase is proportional to frequency. For example, given a delay $T$ the phase is $\phi = -T\omega$, where $\omega = 2\pi f$ is called the radian frequency.

When the area depends on position (e.g., $A(x)$), the phase (and thus the delay) depends on frequency. Fortunately this dependence, between $\phi(f)$ and $A(x)$, is typically small. Since the phase slope defines a delay, which is related to distance by the speed of sound, the phase can provide information about the location of the reflections. Namely the frequency dependent latency can be used to determine where the reflections occur in the ME. A short latency means in the middle ear or TM, while a large latency means

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7i.e., data from Sarah R.
Further more, the phase can tell us even more about the latency of the power absorbed. Since the theory behind the complex reflectance is quite mathematical, it is not presented here. However given the complex reflectance (or equivalently the complex admittance or impedance), the reflections, as a function of depth, is estimated by a well defined procedure.

Ear-canal reflectance and impedance were robustly measured with the use of the multi-cavity technique first developed by Moller [1961], due to Allen [9,10,14], that makes the cavity calibration method more robust to measurement problems. An ear-canal probe that consists of an earphone (called a receiver in the telephone literature) and microphone (aka, a transmitter) is inserted into the ear canal.

### 2.2 False positives and the middle ear

JLM: Remove - out of date JBA: In what way is it out of date? Can you be more specific? Let’s find some clinical person that can discuss this with us, like Linda Hood? I think that what they do is keep testing till they get a pass. They do try to bring the child back at a later time. That does not change the statistics however. Its still very expensive.

A major factor that contributes to the high cost of large-scale (e.g., universal) hearing screening programs is the high rate of false positives. This rate is high because of the inability of current screening methods to distinguish between minor conductive disorders (such as a temporary blockage in the ear canal or ME) and serious inner-ear pathologies (such as a sensorineural hearing loss). The practical consequences of this problem are severe, since the incidence of conductive disorders is roughly 30 times greater than that of inner-ear pathologies in infants [3-5].

Consider, for example, a universal screening program for infants: for every 1,000 infants screened, we expect 2 or 3 to have an inner-ear pathology (0.2%-0.3%) and 50 to 100 to have a conductive disorder (5%-10%). Virtually every infant with a conductive disorder will be positive (fail) the screening test and subsequently require a more extensive evaluation, which is expensive in both time and effort. In addition to high false-positive rates, most hearing screening programs have a narrow scope. The primary objective of these screening programs is to identify children with inner-ear pathologies that can cause negative long-term consequences. Very few screening programs attempt to identify ME pathologies in infants and newborns because of the poor reliability of instruments designed for this purpose and the testing difficulty. A significant weakness of these screening programs exists since chronic conductive disorders, such as otitis media with effusion, also can have serious and long-term negative consequences, thus need to be identified early.

In order for a hearing screening program to be cost-effective, the false-positive rate must be substantially reduced (e.g., reduction by a factor of >30 is required to make false-positive rates negligibly low and referral rates acceptable). The only possible way to achieve this reduction would be with a test that can distinguish between conductive disorders and inner-ear pathologies. This suggests a measure of acoustic power reflectance simultaneously with otoacoustic emissions, so that we can evaluate the status of the ME. Fortunately, instrumentation developed for otoacoustic emission hearing screening can be modified to measure acoustic power reflectance. An instrument that combines these measurements has the potential to simultaneously screen for both ME and inner-
ear pathologies. This would not only reduce the false-positive rate and improve cost-
effectiveness, it would also allow for the identification of a range of different pathologies,
both middle ear and cochlea.

### 2.3 Measurement Technique

The earphone generates the test sound and the microphone then records either a cavity
or ear canal pressure. The acoustic properties of the probe system are then determined
with the use of frequency responses in the four rigid cavities of known impedance. The
ear-canal pressure and the four cavity pressures are then processed to produce an estimate
of the ear-canal power and pressure reflectance.

Etymotic Research, Elk Grove Village, Illinois, developed the ER-10C, a distor-
tion product otoacoustic emission (DPOAE) ear-canal probe used in this investigation.
The frequency-response measurements were obtained with the use of Mimosa Acous-
tic’s HearID (Champaign, Illinois), an automated acoustic measurement system used for
DPOAE measurements. Similar methods of measuring reflectance and impedance have
been developed in recent years. Most of these techniques use a multi-cavity approach
that differ primarily in terms of the size, length and the number of calibration cavities
[11,16].

Reflectance can also be derived from measurements of acoustic impedance with other
methods such as the two microphone method, or the standing-wave tube method [15].
These various methods for measuring the acoustic impedance of the ear have been de-
veloped over the years [17]. Many of these early methods, which do work, have been
shown to be relatively inaccurate. In contrast, the four-cavity method used in this study
is suitable for clinical use and has been shown to be accurate from 0.1 to >8.0 [kHz], as
determined by measurements of standard couplers with known impedance [9] and a pair
of standard acoustic resistors of known resistance [10]. This improvement in accuracy is
mainly due to fewer assumptions. For example, the two microphone method requires a
precise calibration of the microphones, including their relative placement, and the stand-
ing wave tube method requires manual manipulation of the microphone placement. The
main disadvantage of the four cavity method is the need to verify the calibration each
day prior to any measurements, to assure that the ER-10C probe has not changed in
its acoustic properties. As probe technology improves, hopefully this disadvantage will
eventually be removed.

Clinical tympanometric methods have earlier been developed for measuring the re-
ciprocal of impedance (i.e., admittance) at a few frequencies, but the frequency range
of these instruments is limited to around 1 [kHz]. Recently reflectance-tympanometry
has been developed, mainly by Interacoustics. The utility of this composite measure is
presently under investigation. One of the issues with this approach is how to automati-
cally extract clinically useful results, and this has been the subject of on-going research
(Sun, Ear and Hearing articles, 2012-14). Typically tympanometry does not work in in-
fant ears due to the very compliant nature of their ear canal (Margolis et al. [18]; Keefe
and Simmons [19]; Hunter book).

The reflectance measurement protocol in this study, which includes measurements of
acoustic impedance and related acoustic properties of the ear, uses technology that was
initially developed for otoacoustic emission hearing screening [20]. The measurements
have a bandwidth of 0.1 to 6.0 kHz, which is substantially greater than the bandwidths
Verification: This is a procedure that may be used as a check to be sure that the system is working correctly. In its simplest form, one measures the reflectance of a syringe or hard-walled loss-less cavity. In this case $|\Gamma(f)| \approx 1$, which verifies the loss-less condition. This is a very rigorous test, perhaps too rigorous. A deviation from 1 by a few [dB] (1 dB is $\approx 12\%$) in pressure, which is not considered significant. Larger errors are expected above 5 [kHz] due to small phase shifts at higher frequencies. Since an ear is not a rigid cavity, perhaps a more easily interpreted test is to measure in an artificial ear. However such a coupler may not be as handy as a syringe.

The ear-tip can also create errors. If the ear-tip is pushed against the ear canal, creating a small constriction, the phase error may be introduced. Such a block can only be established by replacing the ear-tip back in the canal. At times the ear-tip can become clogged with crumen. In such cases, the solution is to replace the tip with a clean one.

Examples of the absorbance shown on a [dB] axis are given in Fig. 3.

2.4 Normative data sets

Four sets of data are described for illustration and were collected in different laboratories with different power flow measurement equipment. These data are unpublished except for the infant otitis media with effusion (OME) data. The first three data sets for the adult cases (normal, otosclerosis, and perforated ear drum) were collected with the use of the Reflectance Measurement System IV (Mimosa Acoustics, Inc.), which consists of a digital signal processing (DSP) board, a preamplifier, an ER-10C probe, and a calibration cavity set (CC4-V).

The OME data were collected with the Multiple Frequency Impedance and Reflectance Power Flow Analyzer system that consists of a laptop computer with a DSP board, an ER-10C probe, and a similar cavity set (CC4-II). The power flow and reflectance measurements are based upon the four-cavity TM acoustic impedance measurement methods developed by Allen [9]. The data for the three adult ears were collected.
while the subjects were sitting in a quiet office. Repeated measurements were obtained for each test ear whenever possible. The subjects included a female in her 50s for the normal case, a female in her 20s for the case of otosclerosis, and a female in her 30s for the case of the perforated TM. The status of each tested ear was based on a prior clinical diagnosis. DPOAE measurements were obtained from the normal and otosclerotic ears at the time of reflectance measurements. The OME data was collected from a 4-year-old child during a previous study [21]. The subjects in that study were evaluated by certified audiologists.

Audiometric evaluation included air and bone-conducted audiograms and tympanometry at 0.226, 0.678, and 1.0 [kHz] that was measured with the Grason-Stadler, Inc., GSI-33 Middle-Ear Analyzer. All data were collected with subjects consent or parents consent for the minor subject under local institutional review board (IRB) guidelines. The IRB for the normal and the otosclerosis cases was the University of Illinois, Urbana-Champaign, Illinois; the perforated ear drum case was the Carle Foundation Hospital, Urbana, Illinois; and the OME case was the Albert Einstein College of Medicine, Bronx, New York.

The reflectance measurement protocol comprises two steps. The first step is to calibrate the ER-10C ear probe by measuring its Thévenin equivalent parameters, source pressure, and impedance with the same diameter ear tip as for the test ears. This requires measurement of the pressure-frequency responses of the probe for four cavities of different sizes and computation of Thévenin equivalent parameters from the four pressure-frequency responses of the probe. The second step includes measurement of pressure-frequency response from the test ear and the computation of power reflectance with the use of the Thévenin equivalent parameters of the probe. The test ears reflectance and immittance (and other derived parameters) can be computed and plotted from these measurements. In all four cases, the probes and ear tips that were used for reflectance measurements were calibrated immediately before the measurement. The instrumentation that was used in these investigations employs computer-generated stimuli, automated data monitoring, and advanced signal processing for noise and artifact rejection. Data can be collected rapidly and conveniently. Once the system has been calibrated or verified, which takes less than 5 minutes and is typically performed once per day depending on the experimental environment, and the probe has been placed in the subjects ear (23 min), the time for measuring the wideband reflectance of an ear is less than a few seconds.

3 Results

The power reflected from an ear with an ME impairment differs from that of a normal ear. This difference is characteristic of the nature of the ME pathology, hence, the interest in the use of wideband reflectance measurements as a clinical tool [22]. Power reflectance ($|\Gamma(f)|^2$) is only one of several possible acoustic variables that can provide useful information on the status of the ME. Other derived parameters of interest are power absorption ($1 - |\Gamma(f)|^2$), transmittance (power absorption in decibels = $10 \log_{10}(1 - |\Gamma(f)|^2)$), and several forms of acoustic impedance $Z(f)$, such as the reactance, resistance, and impedance magnitude ($|Z(f)|$) (See Table B.1).

Clinically, it is important to determine which of these variables provides the most information about the status of the ME. The figures provide an illustrative comparison
of the acoustic characteristics of the ME: power reflectance, power absorption, transmittance, acoustic resistance, acoustic reactance, and acoustic impedance magnitude. In each case, the variable of interest is shown as a function of frequency, on a logarithmic axis. Data can be obtained over the speech frequencies, from 0.1 to >8 [kHz]. The normalized impedance values are obtained by dividing by the characteristic resistance of the ear canal (ρc/A). The resulting normalized resistance, reactance, and impedance magnitude are dimensionless, since each is the ratio of two quantities with the same units. Normalized values are used to reduce between-subject and between-test variability. The normalization process takes into account between-subject differences in the physical size of the ear (an important consideration when comparing male, female, and juvenile data).

Normal Adult Female  Figure NNN shows data for a female subject in her 50s with clinically normal hearing. The blue curve relates to the left ear, the red curve to the right ear, and the yellow curve to the control based on the standard Buëel & Kjær (B&K) (Norcross, Georgia) 4157 artificial ear coupler [10]. Figure 1(a) shows power reflectance. Power reflectance of the B&K 4157 coupler, which represents the average adult ear, is close to 100 percent at 0.2 [kHz] and decreases monotonically with increasing frequency up to 1 [kHz]. At 1 [kHz], it is close to 40 percent, with a very shallow minimum below 40 percent at 3 [kHz]. Power reflectance then increases above 50 percent with increasing frequency. The data for the two ears are very similar to that of the standard coupler at frequencies up to 3 [kHz]. At higher frequencies, the power reflectances of the two ears differ slightly from each other and have a higher reflectance than that of the coupler. Three repeated measurements were obtained from the right ear and one from the left ear. The equivalent volumes that were computed from the power reflectance curves are 1.56, 1.57, and 1.58 cm³ for the right ear and 1.38 cm³ for the left ear. All the repeated measurements are shown as multiple curves in the corresponding plots with such small variability that they appear as one curve.

The power absorption data (Figure 1(b)) mirror this result. Power absorption for the coupler as well as for the two ears is about 5 percent at 0.2 [kHz] and increases with frequency until it is relatively flat at 1 [kHz]. It then reaches a shallow peak at 3 [kHz]. The average power absorption in this peak region is close to 60 percent. The power absorption for the standard coupler decreases with frequency above 3 [kHz] and falls below 50 percent at frequencies above 4.5 [kHz]. The data for the two ears follow a similar pattern above 3 [kHz] but with a greater decrease in power absorption with increasing frequency than the standard coupler.
The physical interpretation of these data is that at frequencies below 1 [kHz], there is an increasing impedance mismatch at the entrance to the ME and that most of the acoustic power that reaches the TM at these frequencies is reflected back into the ear canal. In contrast, the ME absorbs most of the acoustic power in the frequency region between 1 and 5 [kHz]. This also happens to be the frequency region in which the ear is most sensitive to sound. Because the measurements were made with the use of insert earphones, the free-field pinna response is absent from these measured responses.

Figure 1(c) shows the transmittance, which is the absorbed power transformed to a logarithmic (decibel) scale. Transmittance increases linearly with frequency in the low-frequency region below 1 [kHz] and decreases slightly with frequency in the region above 4 [kHz]. The low-frequency (≤1 [kHz]) slope is approximately 6 dB per octave (20 dB per decade), which corresponds to the impedance of a simple compliance. This figure not only clearly illustrates the absorption of acoustic power by the ME, it also displays a relatively simple picture, in terms of straight-line approximations, of normal power absorption as a function of frequency. In addition, the use of a decibel scale allows for direct comparisons with other relevant data, such as audibility threshold in decibel SPL.

Figure 1(d) shows the normalized acoustic resistance (the real component of acoustic impedance) of the ear canal. The normalized resistance for the coupler is a little over 1.0 in the low frequencies, rises to 1.6 at 1 [kHz], and decreases with frequency to 0.2 at 6 [kHz]. The data for the two ears show a similar pattern except that the normalized resistance at 0.2 [kHz] is on the order of 2 and then falls rapidly with frequency to a value just below 1.0 and remains at this value until 1 [kHz], after which it steadily declines with increasing frequency. The between-ear differences are relatively small. The normalized resistance for the two ears is close to 1.0 over a fairly wide frequency range; for this pair of ears the resistance of the TM is roughly equal to the characteristic impedance of the ear canal (ρc/A = 9 × 10^6 [Rayls]) over this frequency range.

As shown in the normalized reactance data (Figure 1(e)), the reactance of these two ears is not negligible in the mid-frequency region; this means that the impedance of the TM is not entirely resistive and provides only a moderately good match to the characteristic impedance of the ear canal. As a result, about 50 percent of the incident power is reflected back into the ear canal. In terms of transmittance, this corresponds to a loss of 2 to 3 dB in the frequency region from 1 to 4 [kHz]. It should be noted that the reactance of the two ears and that of the standard coupler are in excellent agreement over the entire frequency range.

Over most of the mid-frequency range, the magnitude of the reactance (Figure 1(e)) is larger than that of the resistance (Figure 1(d)), which is approximately equal to the characteristic impedance of the ear canal; this results in relatively efficient power transmission to the ME. Above 3 [kHz], the magnitude of the reactance changes from a stiffness- to a mass-based reactance at which point the power transmission to the ME becomes relatively poor because of the mismatched resistance.

The normalized impedance magnitude is shown in Figure 1(f). This is a concise way of representing impedance in that it summarizes the overall magnitude of the impedance without regard to whether stiffness-based reactance, mass-based reactance, or resistance is the dominant component. Logarithmic scales are used for both the ordinate and abscissa, and the expectation is that in the frequency regions where reactance is the dominant component, the impedance magnitude varies linearly with frequency. In the low frequencies, impedance magnitude is inversely proportional to frequency because the
Figure 2. Adult female with bilateral otosclerosis: Acoustic properties measured in both ears (red for right ear and blue for left) are compared with those of control (standard artificial ear coupler [yellow]) (means in black dashed lines). (a) Power reflectance ($|\Gamma(f)|^2$) in percent, (b) power absorption ($10\log_{10}(1-|R|^2)$) in percent, and (c) transmittance ($10\log_{10}(1-|R|^2)$) in decibels. (d) Normalized resistance, real component $|\Re(Z)/Z_c|$; (e) normalized reactance, imaginary component $|\Im(Z)/Z_c|$; and (f) normalized impedance magnitude ($|Z/Z_c|$) compared with standard artificial ear coupler. Normalized values were obtained by dividing acoustic resistance, reactance, and impedance, respectively, by characteristic impedance of ear canal. Resulting normalized resistance, reactance, and impedance magnitude are dimensionless since each is ratio of two quantities with same units. Frequency ranges from 0.2 to 6.0 [kHz].

3.1 Otosclerosis

Figure 2(a)-(f) shows data for a female adult in her 20s with bilateral otosclerosis. Three repeated measurements were obtained for the right ear and five for the left. All the repeated measurements were plotted as shown by the multiple curves. The means of each ear are shown as black dashed lines. The equivalent volumes were computed for the right ear based on a linear regression of the reactance at low frequencies as 0.8, 0.8, and 0.87 cm$^3$ and for the left ear as 0.93, 0.95, 0.95, 1.06, and 0.98 cm$^3$. These numbers are typical of the equivalent volume test-retest variability. The reflectance data (Figure 2(a)) show that below 0.8 kHz, most of the acoustic power that reaches the ME is reflected back into the ear canal. Between 0.4 and 2 [kHz], the normalized resistance of the young adult otosclerotic ears (Figure 2(d)) is significantly below that of normal ME resistance. Normal ME resistance varies between 1 (at 0.4 and 2 [kHz]) and 1.5 (at 1 [kHz]). This may be explained by the earlier observation that (1) the middle ear is a lossless transmission system and (2) the normal resistance is due to the matched cochlear load. Because of the stiff annular ligament in the otosclerotic ear, a large mismatch in impedance is seen below 2 [kHz]; this causes the incident energy to be reflected back into the ear canal (Figure 2(a)) where it propagates unattenuated. Thus, below 2 [kHz], the reflectance magnitude is close to 1 and the resistance is much smaller than normal. Below 0.4 [kHz], the measurement becomes less accurate as the impedance angle approaches 90° because of the stiffness of the ear canal, which becomes relatively large in the negative direction as frequency decreases (i.e., at 0.3 [kHz] the stiffness is about 10 times greater than the characteristic impedance in the ear canal as shown in Figure 2(e)). The abnormality of these otosclerotic ears is not as clearly evident from the impedance magnitude data (Figure 2(f)).

3.2 Perforated Eardrum

Figure 3 shows data for a female subject with a perforated TM in the right ear (solid red lines for three repeated measurements) and a normal left ear (solid blue line). The mean of the measurements for the right ear is shown by a dashed black line. The perforation
Figure 3. Adult female with perforated TM: Patient has 3-4 mm diameter tympanic membrane perforation in right ear (repeated measurements in red and mean in black dashed lines) compared with relatively normal left ear (blue). Acoustic properties of two ears are compared with those of control (standard artificial ear coupler [yellow]). (a) Power reflectance ($|\Gamma(f)|^2$) in percent, (b) power absorption ($1 - |\Gamma(f)|^2$) in percent, and (c) transmittance ($10 \log_{10}[1 - |\Gamma(f)|^2]$) in decibels. (d) Normalized resistance, real component $\Re(Z/Z_c)$; (e) normalized reactance, imaginary (Im) component $\Im(Z/Z_c)$; and (f) normalized impedance magnitude ($|Z/Z_c|$). Normalized values were obtained by dividing acoustic resistance, reactance, and impedance, respectively, by characteristic impedance of ear canal. Resulting normalized resistance, reactance, and impedance magnitude are dimensionless since each is ratio of two quantities with same units. Frequency ranges from 0.2 to 6.0 [kHz].

In the right TM was a 3 to 4 mm diameter anterior, inferior, central tympanic membrane perforation; size was measured by the ear, nose, and throat doctor at the time of TM repair surgery. The ossicular chain was intact and subjectively mobile. The power reflectance curve (Figure 3(a)) for the left ear is very similar to that of a normal ear up to about 1 [kHz], as represented by the standard coupler measurements. At higher frequencies, the good ear shows a lower reflectance than the standard coupler with a relatively low minimum value at 3.5 [kHz], above which it becomes mass dominated, which causes the reflectance to rise rapidly above this frequency. The ear with the perforated TM (right ear) shows a lower than normal power reflectance in the low-frequency region below 1.5 [kHz]. At higher frequencies, the power reflectance varies over a wide range, although it is consistently below that of the standard coupler. The data become erratic and it is not clear how many minima are relevant in the power reflectance curve. The power absorption data (Figure 3(b)) show a similar high degree of variability. For frequencies below 1 [kHz], the power flow into the ME is substantially greater than that for a normal ear (in terms of transmittance, as specified in decibels). It is not clear in this case that the absorbed power is conducted into the ME. The transmittance curve (Figure 3(c)), in contrast to the power reflectance and power absorption curves, shows relatively little variability over a wide frequency range (0.46 [kHz]). Note that all three curves, power reflectance, power absorption, and transmittance, show a high degree of variability below 0.4 [kHz]. This variability is believed to be the result of external noise that is picked up during the measurement procedure because of the open Eustachian tube of the subject.

The transmittance of the ear with the perforated TM is substantially higher than that of the average normal ear in the frequency region below 1 [kHz], as shown by the curve for the standard coupler. In the intermediate frequency range between 1 and 5 [kHz], the transmittance is still consistently higher than that of the average normal ear but only by a small amount since the transmittance for both the normal and damaged ears is only a few decibels from the maximum transmittance of 0 dB. At frequencies above 5 [kHz], the transmittance of the damaged ear remains high while that of the normal ear (i.e., standard coupler) decreases with increasing frequency. The impedance measurements for the left ear show excellent agreement with the impedance of the standard coupler. The curves for normalized resistance, reactance, and impedance magnitude for the left ear (Figure 3(d)(f)) are remarkably similar to the corresponding curves for the standard coupler over almost the entire frequency range with the only significant differences at either very high or very low frequencies. The data for the damaged ear tell a very dif-
Figure 4. Young child with otitis media with effusion (OME): This child’s acoustic properties (blue) are compared with those of group of 30 young children with normal middle ear function as determined by tympanometry, otoscopic evaluation, and verbal interview. Means of normal ears are plotted as dotted lines in middle of yellow regions, which are mean ± 1 standard deviation. (a) Power reflectance ($|\Gamma(f)|^2$) in percent, (b) power absorption ($1 - |\Gamma(f)|^2$) in percent, and (c) transmittance ($10 \log_{10} [1 - |\Gamma(f)|^2]$) in decibels. (d) Normalized resistance, real component $\Re(Z/Z_c)$; (e) normalized reactance, imaginary component $\Im(Z/Z_c)$; and (f) normalized impedance magnitude ($|Z/Z_c|$). Normalized values were obtained by dividing acoustic resistance, reactance, and impedance, respectively, by characteristic impedance of ear canal. Resulting normalized resistance, reactance, and impedance magnitude are dimensionless since each is ratio of two quantities with same units. Frequency ranges from 0.2 to 6.0 [kHz].

The dip in the normalized impedance magnitude is believed to be a Helmholtz resonance effect between the ME cavity-volume stiffness and the mass of the hole in the TM [2325]. At that frequency, the two reactances are equal and opposite in sign and cancel each other, which results in a sharp dip in the impedance magnitude as resistance is also close to zero in this frequency region.

3.3 Otitis Media with Effusion in a Young Child

Figure 4 shows data obtained from a 4-year-old, male child with OME. Each of the graphs consists of a solid blue line and a broad yellow region of variable width with a dotted line in the middle of the region that indicates a 30-ear mean. The solid blue line represents the data for the OME ear, while the broad yellow region with the dotted line summarizes data for 30 normal ears in the same age group (2.5 to 4 yr). The dotted line at the center of the yellow region represents the average for the 30 normal ears and the boundaries represent the mean ± 1 standard deviation. These data were collected by Wei Wei Lee at Albert Einstein College of Medicine, Bronx, New York, as part of a larger study directed by Judy Gravel [21]. The power reflectance (Figure 4(a)) for the ear with OME is substantially higher than normal, which shows that at almost every frequency, most of the acoustic power that reaches the ME is reflected back into the ear canal. Similarly, power absorption (Figure 4(b)) shows that less than half the incident power is absorbed by the ME. The transmittance for this ear (Figure 4(c)) is more than 6 dB below that of the average normal ear over the entire frequency range. Since 6 dB corresponds to a power ratio of 4:1, the transmittance indicates that less than 25 percent of the incident power is transmitted to the ME over this frequency range.

The observations are not surprising for this case since fluid in the ME restricts move-
ment of the ossicles such that substantially more power is needed to move them. The reduction in transmittance is largely independent of frequency, i.e., transmittance for the ear with OME is consistently below that of the average normal ear by just over 6 dB, except in the region of 4 [kHz] where the difference is less than 6 dB. The data on normalized resistance, reactance, and impedance magnitude also show consistent differences between the ear with OME and the average normal ear.

The normalized resistance (Figure 4(d)) is less than that of the average normal ear over the entire frequency range. While the arithmetic difference is not very large, the ratio is substantial particularly in the frequency region above 2 [kHz] where the normalized resistance for the OME ear is close to zero. As before, this finding is not surprising since the resistance represents that component of ME impedance that absorbs power.

The normalized reactance (Figure 4(e)) of the OME ear is slightly larger in magnitude than that of the average normal ear for frequencies below 2 [kHz]. At higher frequencies, the reactance is approximately the same as that of the normal ear. The normalized impedance magnitude (Figure 4(f)) shows a large difference from that of the normal ear with a marked dip just below 4 [kHz]. All the OME data are significantly different from that of a normal ear. The differences, however, are larger and more noticeable in the three reflectance-based diagrams (Figure 4(a)(c)). Of these, only the transmittance shows the difference between the OME ear and the average normal ear in terms of decibels, which can be related directly to other relevant audiological measures such as differences in hearing level (HL).

4 Discussion

It is useful to recognize how our understanding of audiological measurement has matured based on the available technology. An key example is our heavy dependence on the audiogram as an indicator of abnormal hearing. While it serves us well as a critical measure of hearing dysfunction, it provides almost no diagnostic utility.

The audiogram was developed at a time when the technology of the day (analog electronics) allowed for the frequency response of a system to be measured conveniently. Measurement of the temporal characteristics of hearing was difficult at that time and received little attention in the early development of audiology. The audiogram which is a frequency-based, soon became a routine measure in audiological evaluation. There is no corresponding routine measure of the temporal characteristics of hearing even though modern computer-based technology allows for the convenient measurement of temporal characteristics.

When measuring acoustic power reflectance, we must keep in mind the nature of the quantities that are being measured. In the frequency domain, a pressure wave is a complex quantity and is specified in terms of both amplitude and phase. In mathematical terms, a pressure wave consists of both real and imaginary components. The sum of the squares of the real and imaginary terms is proportional to the power of the plane wave. The ratio of the forward-moving (incident) pressure wave to the reflected (retrograde) pressure wave is called the pressure reflectance $\Gamma(f)$. The square of the pressure reflectance is called the power reflectance $|\Gamma(f)|^2$ and represents the fraction of power reflected by the ear structures (both cochlear and ME). Both the pressure and power reflectances are frequently expressed as percentages rather than fractions. The latency of the reflected power can be determined from the phase of the reflected signal (latency is proportional to the rate of phase change with frequency). Thus, a proper analysis can provide both power magnitude and latency information.

8https://en.wikipedia.org/wiki/Acoustic_impedance
Finally, the complex acoustic impedance of the ear canal (the ME impedance is measured at the microphone location in the ear canal) can be derived from the pressure reflectance and vice-versa. The real component of the complex acoustic impedance is the resistance, while the imaginary component is the reactance (See Table 1).

The data reported in this study are consistent with previous research on the use of reflectance measurements to evaluate ME function. For example, Feeney et al. found abnormal reflectance for otosclerosis (two ears), ossicular discontinuity (two ears), hypermobile tympanic membrane (two ears), perforations of the tympanic membrane (two ears), and a pressurized ME space (two ears) [22]. Hunter found significantly higher reflectance in infants and young children with OME, “poor status” ears, and cleft palates [67].

Others have found similar results [26]. The measurement protocol that was used in this study was developed by Allen [9] and Voss and Allen [10]. In addition to reflectance, this method also provides detailed information on related variables such as acoustic impedance. This article compared several different measurements that can be conveniently obtained with this method and that have application to the clinical assessment of ME function. The measurements compared were a set of three reflectance-based measurements (power reflectance, power absorption, and transmittance) and a set of three normalized impedance-based measurements (acoustic resistance, acoustic reactance, and impedance magnitude). Percent power reflectance was included because of the growing interest in this property of the ear [69,15,25]. Percent power absorption was included since it has an obvious and useful physical interpretation. Transmittance was included because it specifies power absorption on a decibel scale and, in so doing, provides a useful link to other widely used audiological measurements such as HL. Measurements of acoustic impedance were included because of their clinical importance in the assessment of ME function. As described earlier, the impedance-based measurements (resistance, reactance, and impedance magnitude) were normalized by dividing the measured quantities by the acoustic impedance of the ear canal; this reduces between-subject variability by taking into account differences in the physical size of the ear canal and between-test variability by taking into account ambient pressure and temperature differences.

Of the reflectance-based measurements, transmittance appears to be the most useful since it is closely related to the ME transfer function and is specified in decibels. The effect of ME impairment on transmittance can be directly compared with changes in HL. The transmittance of the OME ear in Figure 4(c) was 6 to 10 dB below normal, which was consistent with the elevation in auditory threshold for this ear. To the best of our knowledge, no study has yet made detailed comparisons between transmittance and hearing loss. The normal transmittance curve also has a simple shape which is useful for comparison. The transmittance of the normal ear is approximated quite well by three straight lines: an upward sloping line of 6 dB per octave at frequencies below 1 [kHz], a horizontal line (slope of 0) within 3 dB of the maximum transmittance between 1 and 4 [kHz], and a downward sloping line at higher frequencies (with a slope typically between 6 and 0 dB per octave); this effect is not well understood today. This overall pattern provides a convenient, well-defined reference for testing for abnormal power flow into the ME.

A problem with power reflectance measurements is their relatively high variability in the region from 1 to 3 [kHz] (Figure 1(a)) or 2 to 5 [kHz] (Figure 2(a)). These measurements can highly depend on small experimental errors when the reflectance is small. In
contrast, the transmittance data showed little variability in the frequency regions where reflectance is small because the maximum transmittance was 0 dB. The small variability of transmittance data makes determining when the power flow into the ear is normal easier. The transmittance is more characteristic of hearing threshold measurements than the power reflectance. Another problem with reflectance measurements that does not apply to transmittance is the difficulty of establishing the normal curve in the frequency region of minimum reflectance. Simple averages of many curves can result in a highly biased estimate; this is not an issue with transmittance data. Measurements of power absorption are more easily interpreted than those of power reflectance since it is the absorbed power that determines the sensitivity of the ear. Power absorption measurements, however, are relatively more variable in the region of maximum power absorption because power absorption is directly related (with no change in scale) to power reflectance. The measurements of normalized acoustic resistance and reactance have been shown to be very useful and supplement transmittance data when ME status is being evaluated. Sharp minima in impedance magnitude indicate a special problem. At high frequencies, such minima correspond to mass dominance, which may indicate high frequency problems, while at low frequencies may indicate an TM perforation. Such sharp notches are consistent with most of the energy being reflected by the ME. Notches of this magnitude are of special significance, with the frequency of the notch correlated with the type of abnormality [23-25].

5 Conclusions

In summary, transmittance appears to be the most useful single measure. It shows distinct differences among common ME pathologies that are easy to identify because the transmittance curves are relatively smooth. In addition, the deviation from normal transmittance may be measured in decibels, which specifies the effect of the impairment in audiologically relevant terms. The shape of the normal transmittance curve appears to approximate the ME transfer function, which allows for convenient assessment of abnormal transmittance data. Transmittance, however, does not tell the whole story, and it is advisable that it be used in conjunction with measures of resistance, reactance, and impedance magnitude.

Acknowledgments

Mimosa Acoustics, Inc., developed the equipment used for the measurements we report in this article under a Small Business Innovation Research grant from the National Institute on Deafness and Other Communication Disorders (grant 1 R43 DC03138-01).
A  Intensity and reflectance definitions

760 The intensity for a pure tone at frequency \( f \) is given by

\[
\mathcal{I}(f) = \frac{1}{2} |P(f)U(f)| = 0.5|P(f)|^2/407, \quad (A.1)
\]

where the complex pressure \( P(f) = \rho c U(f) \) at frequency \( f \) and \( |P(f)|^2 \) is its square-magnitude. As an example, if the pressure is 1 [Pascal] (i.e., 94 [dB-SPL]) and the diameter of the ear canal is 7.5 [mm], then the power in the canal is

\[
\mathcal{P} \equiv \mathcal{I}A = 0.5 \cdot 1^2 \pi (3.75 \times 10^{-3})^2/407 = 54.3 \quad [\text{nW}],^9
\]

[9] \( 1 \text{nW} \) is a nano-Watt or \( 10^{-9} [\text{W}] \)

761 Once a reflection occurs, e.g., at the tympanic membrane (TM), the power is split into three parts, the total power \( \mathcal{P}_{\text{canal}} \), the power absorbed by the middle ear (ME) \( \mathcal{P}_{\text{absorbed}} \) and the reflected power \( \mathcal{P}_{\text{reflected}} \). Thus

\[
\mathcal{P}_{\text{canal}} = \mathcal{P}_{\text{absorbed}} - \mathcal{P}_{\text{reflected}} = \mathcal{P}_a \left( 1 - \frac{P_r}{P_a} \right) \quad (A.2)
\]

Henceforth we will use \( \mathcal{P}_c, \mathcal{P}_a, \mathcal{P}_r \) as a proxy for \( \mathcal{P}_{\text{canal}}, \mathcal{P}_{\text{absorbed}}, \mathcal{P}_{\text{reflected}} \).

The canal power reflectance is denoted

\[
|\Gamma_c(f)|^2 = \frac{P_r(f)}{P_a(f)}, \quad (A.3)
\]

762 namely it is the ratio of the power reflected over the power absorbed. The complex pressure reflectance is

\[
\Gamma_c(f) = \frac{P_r(f)}{P_a(f)} = |\Gamma(f)| e^{j\phi(f)} \quad (A.4)
\]

763 The magnitude squared of the pressure reflectance gives the power reflectance. The power-latency, as a function of frequency, is coded in the phase reflectance phase, as the negative of the slope of the phase \( \phi(f) \) with respect to frequency, i.e., \( \tau(f) = -d\phi(f)/df \).^10

B  Summary of Immittance Terminology

764 In Table 1 we summarize most of the terms used in immittance measurements. From the table it should be clear why the topic of immittance is confusing, simply due to the terminology. The use of reflectance, while yet one more term, obviates the need for much of the impedance language. Reflectance is a much more intuitive way of understanding the same physical properties, in an intuitive way.

\[
\begin{align*}
\end{align*}
\]

\(^{10}\)This equation is math, but may safely be ignored.
<table>
<thead>
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<th>Term</th>
<th>Generic Name</th>
<th>Symbol</th>
<th>Units</th>
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</thead>
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<td>Sound Pressure</td>
<td>Pressure</td>
<td>$P(f)$</td>
<td>[Pa]</td>
</tr>
<tr>
<td>Sound Pressure Level</td>
<td>SPL</td>
<td>$10 \log_{10}</td>
<td>P(f)</td>
</tr>
<tr>
<td>Sound Pressure Group Delay</td>
<td>$\tau(f)$</td>
<td>[ms]</td>
<td></td>
</tr>
<tr>
<td>Sound Intensity Level</td>
<td>SIL</td>
<td>$10 \log_{10}</td>
<td>I(f)</td>
</tr>
<tr>
<td>Complex Reflectance</td>
<td>Reflectance</td>
<td>$</td>
<td>\Gamma(f)</td>
</tr>
<tr>
<td>Reflectance Phase</td>
<td>$\phi(f)$</td>
<td>[Rads]</td>
<td></td>
</tr>
<tr>
<td>Reflectance Group Delay</td>
<td>phase slope</td>
<td>$\tau(f)$</td>
<td>[ms]</td>
</tr>
<tr>
<td>Power Reflectance</td>
<td>Reflectance</td>
<td>$</td>
<td>\Gamma(f)</td>
</tr>
<tr>
<td>Power Absorption</td>
<td>Absorbance</td>
<td>$1 -</td>
<td>\Gamma(f)</td>
</tr>
<tr>
<td>Power Transmittance</td>
<td>Transmittance</td>
<td>$10 \log_{10} (1 -</td>
<td>\Gamma</td>
</tr>
<tr>
<td>Admittance Magnitude</td>
<td>Admittance</td>
<td>$</td>
<td>Y(f)</td>
</tr>
<tr>
<td>Impedance Phase</td>
<td>$\phi(f)$</td>
<td>[Rads]</td>
<td></td>
</tr>
<tr>
<td>Impedance (Real Part)</td>
<td>Conductance</td>
<td>$G(f)$</td>
<td>Normalized</td>
</tr>
<tr>
<td>Impedance (imaginary Part)</td>
<td>Susceptance</td>
<td>$S(f)$</td>
<td>Normalized</td>
</tr>
<tr>
<td>Impedance Magnitude</td>
<td>Impedance</td>
<td>$</td>
<td>Z(f)</td>
</tr>
<tr>
<td>Impedance Phase</td>
<td>$\psi(f)$</td>
<td>[Rads]</td>
<td></td>
</tr>
<tr>
<td>Impedance (Real Part)</td>
<td>Resistance</td>
<td>$R(f)$</td>
<td>Normalized</td>
</tr>
<tr>
<td>Impedance (imaginary Part)</td>
<td>Reactance</td>
<td>$X(f)$</td>
<td>Normalized</td>
</tr>
</tbody>
</table>

Table 1: Quantities Relating to Immittance. This table makes clear the inherent complexity of the terminology related to acoustic immittance. This problem is largely historical, and may be largely avoided when working with the reflectance, which has a much simpler interpretation in terms of power, or in the case of complex reflectance, power and latency. The term immittance refers to all the flavors of these measures. It is intended to be nonspecific, perhaps like the generic word *automobile* vs. *Ford*. The impedances have all been normalized by the canal characteristic resistance $r_0 = \rho c / A$. When transforming the impedance into a reflectance, this normalization is necessary, and to reduce the possibility of error in using the wrong value, and in keeping track of the data, we have traditionally worked with the normalized impedance. There is a further advantage beyond computational in that working with the normalized values simplifies the interpretation of the values. This reduces the variability across subjects, since the area is best estimated at the time the data is taken, based on the size of the probe tip used for the measurement.


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These relations summarize pressure, velocity and power. Next we express the wideband TM admittance $Y_{TM}(f)$ and wideband TM reflectance $\Gamma_{TM}(f)$ in terms of these relationships.

While the acoustic power (intensity/area) is the product of the pressure and the velocity, the velocity to pressure ratio defines the complex acoustics admittance

$$Y(f) = \frac{U(f)}{P(f)} = |Y(f)|e^{j\phi(f)} \equiv G(f) + jS(f).$$

Here $G$ and $S$ are the real and imaginary parts of $Y$. These two components are best thought of in terms of the admittance magnitude $|Y|$ and phase $\phi(f)$, rather than in terms of the real and imaginary parts. Typically $Y(f)$ is for single frequencies $f$. 

26