Wideband Tympanometry

What the future holds

Acknowledgments

REFERENCES (Incomplete)
Introduction

A diamond has many facets. Each facet provides a different perspective of the inner beauty. Sound transmission in the ear is a multi-faceted jewel that has garnered much interest by acousticians and clinicians alike. Acoustic power flow at the tympanic membrane (TM) is of particular interest in that it provides a revealing view of sound transmission of sound transmission in the ear canal and middle ear (ME) which, in turn, is mitigated by similar acoustic processes at the boundary between the middle and inner ear.

Historically, the earliest research on sound transmission at the TM was performed by acousticians using tools developed by engineers for circuit analysis. This resulted in the introduction of concepts such as acoustic impedance, resistance and reactance. With time, clinicians recognized the importance of these concepts in evaluating middle ear function and developed clinical methods of measurement from a different perspective, that of admittance which is the reciprocal of impedance. Admittance and impedance are complex quantities and clinicians without the requisite mathematical background had difficulty interpreting the clinical significance of these concepts. In addition, simpler instrumentation covering a narrow, low-frequency range was developed for tympanometry. There was very little overlap between these two facets of sound transmission at the TM. The research literature on the acoustic impedance of the TM has appeared primarily in acoustical journals and the research literature on tympanometry has appeared primarily in clinical journals. Voss and Allen (1994) introduced the concept of reflectance and its relationship to impedance. The concept of reflectance is much easier to grasp without a strong mathematical background. It can also be measured over a wide frequency range.
(up to 10 kHz). These two factors have resulted in a substantial growth in both acoustical and clinical research on acoustic power flow at the TM using reflectance techniques.

This chapter begins with a brief review of the above three facets of acoustic power flow at the TM concluding with a summary of the relative advantages and shortcomings of the perspective provided by each facet.

/H2/ The Impedance Facet

The earliest research on sound transmission in the ear was performed by acousticians, physicists and engineers (with the exception of Hermann Ludwig Ferdinand von Helmholtz who was both a physician and physicist). These early researchers viewed the TM as impedance that opposes the flow of acoustic power into the middle ear (ME) in much the same way that electrical impedance opposes the flow of power in an electrical circuit. The transmission of sound in a tube can also be viewed as being analogous to the transmission of an electrical signal in a transmission line.

A key assumption in analyzing sound transmission in an elastic medium such as air is that the medium is assumed to be made up of infinitesimally small particles and that each particle has a mass and has elastic properties. The term ‘elastic’ refers to the property whereby an applied force can displace the air particles and the work done in displacing the air particles from their quiescent position is stored as compressive energy in the medium. The ‘quiescent position is the steady-state position occupied by an air particle before energy is inserted into the acoustic system. When the applied force is removed, the displaced particles start to return to their initial quiescent positions. The movement of the particles towards their quiescent positions converts the stored compressive energy into kinetic energy.
An air particle returning to its quiescent position will be moving with some velocity and will not suddenly stop moving when it reaches its quiescent position. Its momentum will keep it moving until it displaces an adjacent air particle. The kinetic energy of the moving air particle is then transferred to compressive energy in the newly displaced adjacent air particle. The cycle is then repeated in which the compressive energy in the displaced adjacent air particle is converted to kinetic energy resulting in the displacement of the next adjacent air particle thereby conveying the energy from one air particle to the next. This simplified model of energy transmission in an elastic medium works well in that predictions of energy flow based on this model are consistent with observed energy flow in the audio frequency range. At higher frequencies, other modes of sound propagation come into play.

It is important to distinguish between the two velocities in sound transmission; the velocity of air particles and the velocity with which acoustic energy travels through the medium. The velocity of the air particles is dependent on how the air particles are stimulated by the applied force. The velocity with which acoustic energy travels through an elastic medium is determined by the properties of the medium and is equal to the speed of sound for that medium. The speed of sound in air is about 1100 ft per second. The exact value of the speed of sound in air depends on the temperature and the ambient atmospheric pressure. Variations in temperature and ambient air pressure can have a significant effect on the accuracy of measurement.
The ear canal can be approximated quite well by a hard walled acoustic tube with a fixed cross-sectional area terminating in a membrane with the properties of an intact tympanic membrane (TM). For this condition, it is more convenient to think in terms of pressure rather than force where

\[
\text{Pressure} = \frac{\text{Force}}{\text{Area}} \quad \ldots (2)
\]

A sinusoidal signal is usually used to measure the frequency response of the sound transmission path. Since this energy source is an ongoing stimulus rather than a single pulse of energy, it is more convenient to think in terms of acoustic power flow rather than energy flow where the power delivered to the acoustic system equals the energy delivered per unit time, i.e.,

\[
\text{Power} = \frac{\text{Energy}}{\text{Time}} \quad \ldots (3)
\]

When the diaphragm of the acoustic transducer delivering the sinusoidal signal moves inwards towards the TM, the air particles adjacent to the diaphragm are compressed. Later in the periodic cycle, the diaphragm moves outwards away from the TM causing the air particles to be stretched apart creating a drop in pressure. The cycle in which the instantaneous pressure grows sinusoidally to a peak and then decreases sinusoidally to a minimum (rarefaction) is repeated periodically creating a sinusoidal pressure wave traveling down the ear canal towards the TM. The air particles along this path exhibit similar peaks and valleys in particle velocity that are in opposite phase to the pressure peaks and valleys in. For example, when pressure is at a peak, the displacement of air particles is also at a maximum, but the velocity of the air particles (which is equal to the rate of change in displacement) is zero.
Before moving on to a discussion of acoustic impedance, which involves many subtleties, it is useful to consider how the concept of a complex acoustic impedance was developed starting with the simplest circuit element (a resistor) in a simple DC circuit.

Two fundamental equations serve as the starting point for the discussion that follows:

\[ R = \frac{V}{I} \quad \text{(4a)} \]

and

\[ P = VI \quad \text{(4b)} \]

Where \( V \) = voltage across the conductor,

\( I \) = current flowing through the conductor, and

\( R \) = resistance of the conductor, and

\( P \) = power absorbed by the conductor

Equation 4a is commonly used to define the property known as Resistance. A resistive circuit element will oppose the flow of current through it by an amount proportional to its Resistance, \( R \); i.e., for a fixed voltage across the circuit element, the larger the value of \( R \), the smaller the current flow. This property is evident by rearranging Eqn. 4a: \( I = \frac{V}{R} \).
Eqn. 4b specifies the amount of power absorbed by a resistance R.

Several theorems have been developed specifying the properties of linear DC circuits consisting of resistances. Of particular value are rules derived from these theorems for specifying the equivalent resistance of a complex circuit. Thévenin's theorem, for example, states that any electrical network consisting of linear resistances containing current and voltage sources can be replaced by an equivalent voltage source and an equivalent resistance connected in series.

Another useful set of rules for deriving voltages and currents in a complex circuit are Kirchhoff's voltage and circuit laws. These theorems and associated rules for circuit analysis not only simplify the analysis of DC networks, they also provided the foundation for analyzing, by analogy, much more complex circuits involving alternating currents.

The purpose of this discussion is to draw on the power of analogy to facilitate understanding of complex systems, such as the acoustics of the human ear, by analogy with simpler systems having the same underlying structure. Direct current (DC) circuits are relatively simple to analyze. More difficult alternating current (AC) circuits can be analyzed by analogy with the methods used to solve relatively simple DC circuits since both AC and DC circuits have the same underlying structure subject to an important transformation. The fundamental equations 4a and 4b are essentially the same for both AC and DC circuits provided the phase of the signals is taken into account. Sound transmission in air necessarily involves alternating signals and the more complex problem of analyzing sound transmission in the ear can be simplified substantially by analogy with the methods used to analyze AC circuits. Each step, from analyzing simple DC circuits to analyzing complex AC circuits, to analyzing complex acoustic systems requires a modification of Equations 4a and 4b.
that takes into account the essential difference between AC and DC circuit components and the essential
difference between signal transmission in a uni-dimensional conductor and signal transmission in three-
dimensional space. Once the underlying analogy is understood the increase in difficulty with each step, from
analyzing simple DC circuits to analyzing acoustic power flow at the tympanic Membrane (TM) is relatively
small. In contrast, an analysis from first principles of a complex AC circuit or of acoustic power flow at the TM
is considerably more difficult. For non-mathematicians, the analogies provide a convenient means for
understanding the underlying concepts without doing the math.

/H2/ Reactance: AC analogs of DC Resistance

When alternating currents were introduced in electrical networks, two new circuit elements were introduced;
inductors and capacitors. The analysis of a circuit containing a resistor, inductor and capacitor is not simple. It
requires solving a second-order linear differential equation. However, for the case in which energy inserted into
the circuit by means of an ongoing sinusoidal signal, as commonly done when measuring the frequency
response of an electrical circuit, a much simpler analysis is possible once the circuit has settled down to a
steady-state condition. In the steady-state condition, inductors and capacitors in an AC circuit have a property
known as reactance which is similar to that of resistance, but with an important difference. A reactance is
analogous to a resistance in that it opposes the flow of current. Unlike a resistance, however, a reactance
introduces a 90 degree phase shift between the voltage and current.

There are two types of reactance:

An Inductive Reactance \( X_L = 2\pi f L \) ...............................................................5a

Where \( f \) = frequency in Hz, and
L is the inductance of the inductor in Henrys

The phase of the voltage will lead the current by 90 degrees.

A Capacitive Reactance $X_C = \frac{1}{2\pi f C}$ ................................................................. 5b

Where $f =$frequency in Hz, and

C is capacitance of the capacitor in Farads

The phase of the voltage will lag the current by 90 degrees.

Note that Note that $X_L$ increases linearly with frequency and that $X_C$ decreases linearly with frequency. There will be a frequency at which $X_L$ is equal in magnitude to $X_C$ but opposite in phase so that $X_L$ and $X_C$ cancel each other out at this frequency creating a resonance.

Resistance and reactance both oppose the flow of current in a circuit and the combination of resistance and reactance in opposing current flow in a circuit defines the impedence of the circuit. Because of the phase difference between voltage and current in a reactance (as opposed to zero phase difference in a resistance) it is necessary to use vector addition to obtain the impedance. Using the resistance vector as a reference, the magnitude of the resistance is shown horizontally. The vector representing the reactance is at right angles to the resistance vector. The sum of the two vectors represents the impedance.

A DIAGRAM REPRESENTING THE VECTOR SUM WOULD BE USEFUL

Expressed mathematically,

Impedance $Z = R + j X$

................................................................. 6
Where $R$ is a real number and $X$ is an imaginary number

The fundamental equations 4a and 4b are restated below for the case of AC signals

Impedance $Z = \frac{V}{I}$ ................................................................. 7a

and

$P = V I \cos(\theta)$................................................................. 7b

Where $V =$ voltage across the circuit element,

$I =$ current flowing through the circuit element,

$Z =$ impedance $= \text{Resistance } R + \text{Reactance } X,$

$P =$ power absorbed by the conductor, and

$\theta =$ phase angle between $V$ and $I$

Bear in mind that the above equations apply for the case of steady-state sinusoidal AC and that $V$, $I$, $Z$, and $X$ are complex quantities and must be specified in terms of both amplitude and phase. Equations 7a and 8a are quite general and apply to both AC and DC circuit elements.

For example, voltage and current are in phase in a resistance in a DC circuit; i.e., the phase angle $\theta = 0^\circ$, \( \cos(0^\circ) = 1 \) and Eqn. 7b reduces Eqn. 4b for a DC circuit.
Equations 7a and 7b can be used to analyze AC circuits using the theorems and rules developed for DC circuits, such as Thévenin's theorem, the derivation of equivalent impedance, and Kirchhoff's voltage and circuit laws, and the derivation of equivalent impedance.

/H2/ Acoustic Signal Analysis: Analogies with AC Signal Analysis

This section extends the analogy with AC electrical circuits to the analysis of sound transmission in the ear. The following are key analogies:

Pressure is analogous to voltage in an AC electric circuit.

Velocity of the air particles conveying the pressure wave is the acoustic analog of current.

Bear in mind that the analogy holds for sinusoidal signals and that the phase relationships among the variables is an important factor in the transmission of sound in air. The displacement of air particles by the applied pressure has the same phase as the pressure wave; i.e., when the pressure wave is at a maximum, the displacement of air particles is also at a maximum. The phase of the volume velocity, however, leads the phase of the pressure wave by 90°. This is because velocity is the time derivative of displacement. The displacement of an air particle about its quiescent position is sinusoidal; i.e., the displacement is specified by a sine wave and the time derivative of a sine wave is a cosine wave, which is equal to a sine wave with a 90 degree phase lead.

Substituting pressure for voltage and particle velocity for current in Eqn. 7a yields the following definition of acoustic impedance

Impedance $Z = \frac{V}{I} = \frac{\text{(pressure)}}{\text{(particle velocity)}}$…………………………………… 8a

The acoustic impedance shown in Eqn. 8a is defined as the **specific characteristic impedance**
It is a property of the medium. For the case of air it can be shown that the

specific characteristic impedance of air = \( \rho c = 407 \) [Rayls]

where \( c \) is the speed of sound, and

Consider now the transmission of sound in the ear canal. In this case it is convenient to think of the air in the canal being made up of a series of discs of air infinitely narrow width. Each disc having a mass and elasticity. For this acoustic system the analog of current is the particle velocity multiplied by the cross-sectional area of the ear canal. This product has the dimensions of a volume and is referred to as the volume velocity. This term has been major source of confusion for non-mathematicians. For some, it may be useful to bear in mind that volume velocity is essentially particle velocity multiplied by a relevant constant (“A rose by any other name…….”)

Substituting pressure for voltage and volume velocity for current in Eqn. 7a yields the following definition of acoustic impedance

\[
\text{Impedance } Z = \frac{V}{I} = \frac{\text{pressure}}{\text{volume velocity}}
\]

The acoustic impedance shown in Eqn. 8b is defined as the **characteristic impedance of the ear canal**

Note that the characteristic impedance of the ear canal is equal to the specific characteristic impedance of air divided by the cross-sectional area of the ear canal.
ADD A BRIEF SUMMARY OF THE ADVANTAGES AND LIMITATIONS OF THE IMPEDANCE FACET

/H2/ The Reflectance Facet

The introduction in the current draft can be inserted here.
When an acoustic pressure wave \( P(f) \) travels down the ear canal toward the tympanic membrane (TM), the frequency-dependent acoustic power \( P(f) \) [W], 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. Here we denote reflectance by the upper case Greek letter "gamma" \( \Gamma(f) \).

In this chapter we will discuss methods of noninvasive ear canal acoustic measurements to diagnose the function of the middle ear and cochlear. These methods have been referred to by various names, including Wideband Acoustic Energy reflectance, Power reflectance, Wideband acoustic impedance and admittance and most recently Wideband acoustic immittance (WAI).

Wideband acoustic impedance/admittance (aka, immittance) is closely related to wide-band acoustic reflectance, the two being mathematically equivalent. In this chapter we shall concentrate on acoustic reflectance, as it is more easily understood than immittance, as explained below.

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1 For clarity all units will be displayed in square brackets using the international standard SI system [kgm, meters, seconds, ...].

2 All abbreviations are provided on the title page.

3 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.
Absorbed and reflected waves

The concept behind wideband acoustic reflectance is to decompose the acoustic pressure wave into absorbed and reflected components. When sound is presented to the ear canal via a small loudspeaker, the pressure wave travels down the ear canal (toward the TM), where a small portion of it is reflected back toward the source. It is helpful to define a notation for these pressure waves. The absorbed pressure $P_+(f, x) \,[\text{Pa}]$ (having frequency $f \,[\text{Hz}]$ and location $x \,[\text{m}]$) is the forward traveling wave, denoted with a subscript $\pm$. The pressure has units of Pascals [Pa]. It may be helpful to think of pressure as the acoustic voltage. The pressure (like voltage) is a scalar (it has no direction). Any change in the pressure results in a force. The reflected backward traveling (retrograde) pressure is denoted $P_-(f)$. At each point in the canal these two pressure add giving the total pressure $P(f, x)$, at location $x$ along the canal

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

Let $x = 0$ label the speaker location and $x = L$ the location of the TM.

A second variable is the acoustic volume velocity $U(f, x)$, defined as the total velocity of the wave at any point $x$ in the canal. As with the pressure, it may be decomposed into a forward $U_+(f, x)$ and a reverse $U_-(f, x)$ traveling portions, as

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

The velocity, being a flow, is a vector (it has a direction), which accounts for the change in sign of Eq. 2.

At every point $x$ along the canal, the ratios of the pressures $P_\pm$ and velocities $U_\pm$ are equal to the acoustic resistance $r_0(x)$ of the ear canal. Due to conservation of energy $r_0(x)$ resistance must always be positive and real, defined as
Eq. (3).

Here \( \rho \approx 1.2 [\text{kg/m}^3] \) is the density of air, \( c \approx 343 [\text{m/s}] \) is the speed of sound (in air) and \( A(x) \) is the canal area at location \( x \) along the ear canal. The product of the density and sound speed is called the specific acoustic resistance \( \rho c = 412 \) [Rayls]. The average diameter of the canal is about 0.75 [cm], with average area of \( 44.2 \times 10^{-6} [\text{m}^2] \)

Reorganizing this relation we obtain the definition of the complex acoustic reflectance

\[ \Gamma(f, x) \]

as a ratio of reflected to absorbed pressure (or velocity). Since \( \Gamma(f, x) \) is complex, it may be written either as the sum of real and imaginary parts, or in terms of its magnitude and phase (Eq. A.4, Appendix A).

The power of the complex reflectance over the complex immittance is that the magnitude reflectance (\( |\Gamma(f, x)e^{j\phi(f,x)}| = |\Gamma(f, x)| \)) describes the relative absorbed power as a function of frequency, while the phase \( \phi(f, x) \) codifies the latency of the absorbed power (e.g., depth of the reflected wave). To date, these properties of reflectance have rendered it more intuitive than impedance, when formulating diagnoses of ME pathologies. When the canal has a constant area and is loss-less, \( |\Gamma(f)| \) is independent of the probe (earphone) insertion depth (Voss, 2013, ARO & AAS). Part of the problem with immittance is the obtuse terminology, as

4 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 \( j\phi \) (phase).

Here \( e^{j\phi} = \cos(\phi) + j \sin(\phi) \) is called Euler's formula \( (j = \sqrt{-1}) \).
outlined in Table B.1 (Appendix B). Additionally the transformation from complex reflectance to complex impedance is complicated [Voss and Allen, 1994]. Specifically the complex impedance may be computed from
the complex reflectance by the formula

\[ 1 + \Gamma(f, x) \]

\[ Z(f, x) = r_0 \left( 1 - \Gamma(f, x) \right). \]  

The complex admittance is given by \( Y(f, x) = 1/Z(x, f) \). Other than for the simplest of cases (e.g., \(|\Gamma| = 0\)), this transformation is mathematically challenging.

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, compared to immittance, we shall focus on the complex reflectance in this report.

Calibration of Sound Pressure Level

Figure 1: Forward pressure level normalization factor which corrects the ear canal standing wave.

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 (that part of the ear canal...
between the earphone and the TM. When the area of the canal is constant, the TM reflectance $\Gamma_{tm}(f, L)$ is related to the measured microphone reflectance $\Gamma_m(f, x = 0)$ by the relation

$$\Gamma_m(f, 0) = \Gamma_{tm}(f, L) e^{i2\pi f \cdot 2L/c} \quad \text{Eq. (6)}$$
where $L$ is the distance from microphone to TM ($c = 343$ [m/s] is the speed of sound).\textsuperscript{77} Simply stated there is a round trip delay of $2L/c$ between the measurement point and the TM. Given the measured $\Gamma(f, 0)$ and the canal length $L$, one may compute $\Gamma_{tm}(f, L)$, and thus the TM immittance. When the area of the canal depends on position, the effective length depends on frequency (i.e., $L(f)$), and the relationships become more complicated.

If one can estimate $L(f)$, then one may estimate $\Gamma_{tm}(f, L)$ from the measured $\Gamma_m(f)$. An procedure for estimating $L$ from $\Gamma_m(f, x)$, which allows for a variable canal area, was reported by Robinson and Allen (HR, 2013) and by Rasetshwane and Neely (JASA, 30(5), p. 3873, 2011). For most clinical applications $\Gamma_{tm}(f, L)$ 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.

\textbf{Figure 2: Forward pressure level normalization factor which corrects for 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 of the null critically depends on the round trip delay from the source back to the microphone (Eq. 6). This delay is different for each ear, as it depends on the insertion depth of the probe and the geometry of the canal and TM.}

/H2/ Clinical utility of the complex reflectance

Next we discuss some of the main applications of the complex reflectance method. The first is the calculation of the \textit{forward pressure level} which allows for the individualized calibration of the signal in the ear canal so that ear canal standing waves are removed and the pressure delivered to the cochlea is approximately constant. A second application is a method to remove the effect of the \textit{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 middle ear pathologies.

The research clinician would most like to know the transmission properties of the ME. Specifically, is the ME clinically normal? If not, in what way is it abnormal? There are many 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 quantify and then identify (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 attempt to improve the calibration of the earphone by placing a microphone in the ear canal, to account for variation in the subject’s ME impedance. This method is intended to replace traditional, but less reliable, artificial ear calibrations, which do not account for these subject-dependent ME impedance variations. In principle, measuring the actual ME pressure rather than using an artificial ear (coupler) to predict it, seems like the ideal approach. However there are several serious and unrecognized problems that need to be addressed. The most obvious of these is the effect of **canal standing waves** [Siegel, JASA 1994]. A standing wave is created when the forward and backward pressure are out of phase, thus nearly canceling. This phase shift is due to the round-trip delay from the source $x = 0$ to the TM $x = L$.

*When an 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 $\Gamma_m(f, L)$, in a frequency-dependent manner. This results in a backward propagating wave $P_-(f, x) = \Gamma(f, x)P_+(f)$. All the acoustic characteristics of the ME, beginning at the TM, are captured by $\Gamma_m(f, 0)$ at the microphone ($x = 0$). When the microphone measurement location is not at the TM, there is a phase factor given by the round-trip delay $2L/c$ between the TM and the microphone location, which causes a standing wave. When this phase equals $180^\circ$, the pressure has a deep null, as seen in Fig. 2.

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. For a review of this problem see Gorga and Neely (1998), Souza (2014) and Keefe et al. (2011?).

The nature of these standing waves is shown in Fig. 2. Based on the deep nulls seen in the figure, clearly it 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

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5 Within the clinical audiology community it is well known that imittance can be a difficult concept to master.

6 This transformation is related to a Möbius, or bilinear, transformation.

7 Estimating $\Gamma_m(f, 0)$ requires a special calibration of the earphone, known in the Engineering literature as a Thévenen calibration (http://en.wikipedia.org/wiki/Thevenin’s_theorem). While quite important, this measure is beyond the scope of the present discussion.
re-normalized to be constant. Real-ear calibrations have the unintended effect of putting a large peak at the standing wave frequency.

Thus the frequency of the resulting null in microphone pressure depends on the probe insertion depth. If one corrects for the microphone pressure (make it constant), then a large peak is introduced at the frequency of the null, resulting in a very serious error in the calibration. Standing waves need to be dealt with or else real ear calibrations are not practical above a few kHz.

It is now recognized that the forward pressure level (FPL) should be the normalized measure (Souza, Dhar, Neely, Siegel (2014) (9 methods; JASA)). However, other than some research systems, FPL has not been integrated into clinical equipment at the time of this writing.

This calibration procedure is a variant on the so-called reference equivalent sound pressure level (RETSPL) method, where the voltage on the transducer is varied to main- tain the pressure at the TM constant in an artificial ear. Here we coin the term reference equivalent forward pressure level (RETFPL). Given the microphone pressure $P_m(f, 0)$ and its corresponding complex reflectance $\Gamma_m(f, 0)$, the forward pressure $P_+(f, 0)$ can be computed. Then the receiver voltage is varied so that $P_+(f)$ is constant at the desired level.

One may determine $P_+(f, x = 0)$ as follows:

$$P_m(f) = P_+(f) + P_-(f)$$

$$= P_+(f) \left(1 + \frac{1}{P_+} \right)$$

$$= P_+(f)(1 + \Gamma_m(f)).$$

Solving for $P_+(f, 0)$ gives
\[ P_+(f) = \frac{P_m(f)}{1 + \Gamma(f)}. \]
Since the microphone pressure and the complex reflectance are both known, one may solve for the absorbed (i.e., forward) pressure by the above relation. Without knowledge of the individual’s complex reflectance $\Gamma(f)$, the standing wave many not be precisely removed.

Figure 2 shows the magnitude of the normalization factor $(1 + \Gamma(f))/2$ for 10 human ears, taken from Voss and Allen (1994). As the frequency is increased above 3 [kHz], the phase of $\Gamma(f)$ plays a very important role, as it can result in ear canal standing waves, as seen at 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. 2, have been computed from the complex reflectance using the relation

$$P_-(f) = P_+(f)$$

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 factor of 2 (i.e., 6 dB$^8$) is optional, depending on the interpretation of what is desired at low frequencies.

To experimentally verify this formula, consider the following experiment. If we measure the pressure at the end of a rigid cavity, and normalize $P_+(f)$ to be a constant (say 1 [Pa]), then at the end of the cavity, $\Gamma(f) = 1$ (since the cavity is rigid). Thus $P_-(f) = P_+(f)$

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$^8$ This factor is known in the literature as the missing 6 dB.
It follows that the pressure at the end of the cavity is 2 since \( P = P_+ + P_- = 1 + 1 = 2 \) [Pa].

**TM Acoustic admittance:** From the clinical diagnostic point of view, the most important quantity is the acoustic admittance of the TM \( Y_{tm}(f, L) \), as a function of frequency. This is what tympanometry attempts to estimate (Rabinowitz, 1981; Shanks et al., 1981). 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, \( \Gamma_{tm}(f, L) = \frac{P_-(f, L)}{P_+(f, L)} \). As previously stated, given \( \Gamma_{tm}(f, L) \), the complex TM admittance and impedance are easily determined (Eq. 5). The proposal then is to Thévenin calibrate the ear canal probe. 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 complex \( \Gamma_{tm}(f, L) \). This is the starting point for the ME diagnosis.
Overview of Middle Ear Impedance and Reflectance

The middle ear is remarkable for its functionality in converting airborne sound to acoustic vibration of the TM. This vibration drives the ossicles, which transmit the acoustic signal to the annular ligament of the oval window, which in turn conveys the signal to the fluid-filled cochlea. Following this are the ossicles, consisting of the masses of the malleus, incus and stapes, shown here as electrical inductors. Each of these masses is the ligament which is represented as a spring. As in the cochlear circuit, this ligament is nonlinear since it changes its compliance when force is applied via the stapedius muscle. Finally, the cochlear compliance is represented as a resistance. This is also nonlinear in that it generates acoustic otoacoustic emissions, which are low-level retrograde signals (i.e., OAEs), close to the threshold of hearing for the normal ear. 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 ME transmission lines may not be ignored and the full model must be considered.

In the normal ME sounds are 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 normal middle ear is a cascade of transmission lines that are approximately matched, having relatively little loss and reflection along the pathway from the ear canal to the inner ear (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 [dB-SPL]. This intensity corresponds to an average motion of the stapes of \( \approx 8 \) [pm] (i.e., \( \approx 1/3 \) the radius of the Hydrogen atom (25 [pm]).

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 via the ME and then evoked retrograde otoacoustic emissions which travel back out through the ME to the ear canal. Both directions of propagation are subject to attenuation from an abnormal ME.

Due to the reciprocal nature of the middle ear, it is also true that little power is lost for reverse traveling signals, which have similar loss characteristic as the forward traveling waves (Allen and Fahey, 1992). It follows that small vibrations generated in the cochlea by nonlinear motions of the outer hair cells can be measured in the ear canal. Measurements of these signals, known as otoacoustic emissions (OAEs), are an important tool for studying the function of the inner ear. This has led to the development of powerful new techniques for the objective assessment of sensorineural hearing loss, leading to cost-effective methods for world-wide hearing screening 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\footnote{These signals were discovered by David Kemp and Duck on Kim, as first reported in the early 1970s.}
mechanism protects the inner ear from intense air-born 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 (TT references: Aron, M., Floyd, D., Bance, M. (2015). Voluntary eardrum movement: A marker for tensor tympani contraction? Otol Neurotol, 36(2), 373-381.; Bance, M., Makki, F. M., Garland, P., et al. (2013). Effects of tensor tympani muscle contraction on the middle ear and markers of a contracted muscle. Laryngoscope, 123(4), 1021-1027).

/H2/ Impedance Discontinuities cause Reflected Acoustic Power

The term “middle ear impedance” typically means 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. 3). 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; Lynch TJ 3rd, Peake WT, Rosowski JJ., 1994) 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 and cochlea (Puria and Allen, 1998). 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 consists of several stiffness components and several mass components (see Fig. 3) 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 relatively 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 (Lynch et al, 1994).
Between 1 and 5 [kHz] the resistance and reactance are of comparable magnitude, whereas at higher frequencies the mass reactance is expected to be at least several times greater than the cochlear resistance.

/H1/ Methods

/H2/ General properties of reflectance and admittance

Definitions of acoustic variables: The acoustic intensity of sound in the ear canal is
\[ I(t) = p(t)u(t) \text{[W/m}^2\text{]}, \]
where \( p(t) \) is the pressure in Pascals [Pa] and \( u(t) \) is the volume velocity \([\text{m}^3/\text{s}]\), as functions of time \( t \). The acoustic power \( P \) is the Acoustic intensity \( I \) times the cross-sectional area \( A \) \([\text{m}^2]\) of the ear canal, namely \( P(t) = A(x) I(t) \text{[W]} \). In general the area varies along the length of the canal \( x \) \([\text{m}]\). The acoustic particle velocity \([\text{m/s}]\) times the canal area \( A(x) \) is called the volume velocity \([\text{m}^3/\text{s}]\).

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 are in terms of a pure tone of frequency \( f \).

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.\(^{10}\) Namely the frequency dependent latency can be used to determine where the reflections occur in the ME. A short

\(^{10}\) i.e., data from Sarah R.; ??Neely and Merchant???
latency means in the middle ear or TM, while a large latency means deep in the cochlea.  

Examples of this relation will be provided in later sections.

The reflectance phase can tell us even more about the latency of the power absorbed. Given the complex reflectance (or equivalently the complex admittance or impedance), the reflections, as a function of depth, may be directly estimated.

Ear-canal reflectance and impedance were robustly measured with the use of the multi-cavity technique first developed by Moller (1961), and extended by Allen (1986), Voss and Allen (1994) and Puria and Allen (1998), that makes the cavity calibration method more robust to measurement error.

/ H2 / 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? Lets 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. It is 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
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.

/H2/ 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 distortion product otoacoustic emission (DPOAE) ear-canal probe used in this investigation. The frequency-response measurements were obtained with the use of Mimosa Acoustic’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 developed 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 standing 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 reciprocal 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 automatically 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 infant 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 of most clinical instruments currently in use for assessing ME function (e.g., instruments for measuring acoustic immittance). Unlike tympanometry, the system described here does not vary the static pressure that is applied to the TM. While such a measurement is useful for determining the static pressure behind the TM, it greatly complicates acoustic power assessment.

**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 cerumen. 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. 4.

/H1/ Results

/H1/ Discussion

It is useful to recognize how our understanding of audiological measurement has matured based on the available technology. A 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.
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.
Clinical applications

The middle ear is a complex mechanism with many components. There are many possible disorders of the middle ear, including fluid in the ear, ossification of the bony structures, discontinuities of the ossicular chain, and perforation of the eardrum, as well as abnormalities of the membranes, ligaments, and supporting structures. Since the middle ear is involved in virtually every test of hearing, it is critical to ascertain the middle-ear status at the outset of any audiological evaluation, and, in the case of abnormal middle-ear function, pinpoint the cause to enable an appropriate intervention. Wideband reflectance measurements evaluate the middle ear over a wide frequency range (0.2 to 6 kHz, or higher), allowing clinicians to make more nuanced interpretations of hearing health. For no extra effort, the forward pressure level can be used to improve stimulus calibration, and increase test validity and reliability. Although complex pressure reflectance offers a more complete picture, to date, clinical researchers have focused on power reflectance and absorbance.

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. Scientific models are used to simulate and understand the underlining principles of the mechanism in normal and abnormal conditions. Combining the outcome of these methods allows us to derive workable diagnostic criteria. In the discussions below, different studies demonstrate different approaches to establish criteria. In this section, many of the described findings can be used by clinicians and researchers with clinical WAI equipment. In some cases the diagnostic criteria are built in, and in other cases, the clinician or researcher may need to derive the result from the exported data. The clinical utility of WAI is undergoing intensive research, and the outcomes are rapidly advancing into clinical applications. Here we provide an overview of clinical norms, various ME pathologies, and sources of variability and other challenges in clinical research.
A single WAI measurement produces a wealth of information. High-resolution analysis provides frequency resolution in the order of 20 Hz over at least a 0.2 to 6 kHz range. Multiple quantities may be derived, as described earlier, from power reflectance and absorbance to a range of other immittance values. 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 reflectance and absorbance level across frequency can describe frequency-dependent behavior using a smaller set of parameters (Hunter, Feeney, Lapsley Miller, Jeng, & Bohning, 2010). One-third, one-half, and whole octave bands are frequently used, and fit nicely with other audiological tests, while still capturing the shape of the power reflectance curve. It is not typically useful to take the average across the entire curve, because frequency dependent behavior will be 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 reflectance or absorbance curve may be modeled 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. *Reflectance Phase Analysis.* Of particular interest is separating the complex (magnitude and phase) pressure reflectance into ear canal and tympanic membrane components, as this retains time-delay information at the tympanic membrane. A number of investigators have proposed methods to remove ear canal phase effects.

\[H2\] Norms

For clinical use, what first needs to be established is 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 a group of highly-screened normal ears for a specific demographic group. The type and degree of screening for defining “normal” varies across studies, but includes audiological history and audiological tests such as AABR, OAE, tympanometry, surgical discovery, and pneumatic otoscopy. Different screening tests and criteria can be a reason for differences between norms across studies. Norms can be expressed in various ways, including percentile ranges and means with standard deviations.
Figure 5. Example norms for four age groups, all plotted as 10-90th percentiles (gray region), median (solid black line), and 5 randomly chosen examples (dotted lines). 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. (2010); Infant norms are from Hunter, Tubbaugh, Jackson, and Propes (2008); Child norms (1/6 octave average bands) are from Beers, Shahnaz, Westerberg, and Kozak (2010); Adult norms are from Rosowski et al. (2012).

For WAI, the demographic that has the greatest effect on the middle ear is age, followed to a much lesser extent by ethnicity. Error: Reference source not found shows norms for four key age groups, replotted from their original studies in consistent units (median and 10th-90th percentile range) to aid comparison. Sex and ear differences are much smaller, and typically not observed. 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 more larger-N normative studies are needed with 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.
Taking a norm may obscure individual patterns across frequency and then 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 Error: Reference source not found with randomly-selected examples.

**Newborn and infant norms:** The human ear undergoes significant maturation in the first 12 months of life (as summarized in Kei, Sanford, Prieve, & Hunter, 2013). They found that the largest changes in WAI norms occur between birth and 6 months of age, 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, and thus less absorptive (Kei et al., 2013). With age, changes in ossicle bone density, along with the loss of mesenchyme and other middle ear fluids, leads to decreased mass in the middle ear system (Aithal et al., 2014b; Kei et al., 2013). This leads to an increase in the ME absorbance at high frequencies, as the lower mass causes less of the signal to be reflected. This rapid maturation means norms are potentially needed for many age bands.

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. In Absorbance, the norms typically show unreliable results below 1 kHz due to environmental noise and ear-tip leaks, high absorbance at 1-2 kHz (higher than in older ears), decreasing at 3-4 kHz, and rising again at 6 kHz (which is also not seen in norms for older ears). Error: Reference source not found 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. For instance, the ear canal contribution to WAI varies across measurement systems and testers, as some types of probes are inserted more deeply into the ear canal than others.

Aithal et al. (2013) correctly pointed out that using just 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 ME measurements. Whether this is an issue or not depends on the purpose. If the aim is to understand the normal infant ME, it is important. But if the purpose of is to assess infant inner-ear status (as in UNHS programs) any temporary ME 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, difference in ethnicity were found by Aithal, Kei, and Driscoll (2014a) where Australian Aboriginal infants had lower wideband absorbance than Australian Caucasian infants. This could be of clinical importance due to the high otitis media with effusion (OME) prevalence among Aboriginal children.

For slightly older infants, Hunter et al. (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. Error: Reference source not found shows the challenges involved in testing this age group, with evidence of noise and leaky probe fits at the lower frequencies, which cause a increase in the absorbance level (or decrease in reflectance).
The study with the largest number of normative subjects for young children is Beers et al. (2010). They tested wideband reflectance in 78 children (144 ears) age 5-7 with normal middle-ear function for comparison to those with OME. Error: Reference source not found shows less variability in part because these data are averaged in 1/6 octave bands, but also because older children are quieter. Of interest is their comparison between Caucasian and Chinese children’s ears, where significant differences were found at 2 and 6 kHz. As we’ll see, 2 kHz is an important frequency 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 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 reflectance curve was similar to previous studies (Error: Reference source not found). Of interest is the parameterization of the absorbance curve. In log-log coordinates, the curve can be modeled with 3 straight lines (Allen, Jeng, & Levitt, 2005; Rosowski et al., 2012). Below 1 kHz, 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/reflectance curves and deriving parametric values to characterize them could aid in clinical decision making. For instance, changes in slopes or frequency of intercepts, or deviation from a straight line may be indicative of abnormal ME performance.

**Summary:** As discussed by Shahnaz, Feeney, and Schairer (2013) norms age, gender,
studies suggest where larger studies should look. However, we need larger N studies to understand how reflectance behaves statistically in the population, for each pathology and demographic. These results must then be combined and reduced to specific criteria for decision making. Needed are larger N studies for specific pathologies, as well as across a range of confusible pathologies to reveal differential diagnoses. Fortunately, a number of such studies are occurring, and we will summarize some here.

As discussed by Voss (2012), although the clinical data shows systematic differences in the presence of pathology due to physiological changes, the high degree of variability across normal subjects can mean that it is not always possible to clearly differentiate normal from diseased ears. In most cases, no test on its own will provide a definitive diagnosis. However, the addition of WAI to the audiologist’s arsenal provides new information that improves on our current abilities to detect and diagnose pathology.

### WAI in universal newborn hearing screening programs

The goal of universal newborn hearing screening (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. Hunter et al. (2010) showed why rescreening OAEs after a delay was often successful –
middle-ear reflectance tends to decrease 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 WAI reflectance was vastly superior to tympanometry in predicting which ears would show low OAE levels due to middle-ear dysfunction. Adding WAI 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, high reflectance/low absorbance is also cause for follow-up for middle-ear dysfunction like otitis media.

**Call out box – 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 6 with real examples from real babies in the Hunter et al. (2010) study. This study showed that the reflectance/absorbance around 2 kHz was the best predictor for whether the DPOAE test passed (DPOAE levels at 3 or 4 out of 4 frequencies are normal) or not (DPOAEs at 2 or more out of 4 frequencies are abnormally low). So although absorbance is plotted from 1 to 6 kHz, pay particular attention to the 2 kHz region. Reflectance below 1 kHz is not plotted because in babies this region is often noisy and therefore is less diagnostic (Hunter et al., 2010).

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

How could WAI be used in UNHS programs? Specific guidelines are still in development, but potentially WAI can be used to enable smarter timing for rescreenings and follow-ups. For instance, in the examples in Figure 6, the following courses of action may be appropriate:

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 is commonly seen in newborn hearing screening programs, and causes 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 it 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 OME/CHL in infants and children

Identifying conductive hearing loss (CHL) in young infants can be difficult to do 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 ABR tests along with tympanometry and wideband reflectance (43 ears had normal hearing and 17 ears had CHL, determined from the air- and bone-conducted ABR thresholds). Prieve et al. found that wideband reflectance between 800 and 3000 Hz was higher in CHL ears compared with normal ears. This is consistent with increased stiffness in the middle ear. Prieve et al. found that a criterion for reflectance greater than 69% in the one-third octave band around 1600 Hz produced the highest likelihood ratio for CHL, compared to other reflectance bands and compared to various quantities derived from multi-frequency tympanometry (at 226, 678, and 1000 Hz). These results indicate the frequency range most sensitive to CHL in infants, and that WAI is a suitable replacement for tympanometry in this age group. But with only 17 ears with CHL, a larger study is needed to
more accurately determine appropriate criteria, as well as investigating if age-specific criteria are needed due to rapidly changing middle-ear physiology in the first year of life.

In children, otitis media is the most common reason for CHL. Middle-ear effusion (MEE) and negative middle-ear pressure (NMEP) tends to stiffen the middle ear and thereby increase the amount of energy reflected, particularly around 1-3 kHz, compared to normal ears. This increased reflectance can be used to identify MEE. Hunter et al. (2008) points out that tympanometry is unreliable in very young infants, and that it can produce normal results in the presence of MEE. It is important to be able to diagnose MEE in this population. They evaluated wideband reflectance for its ability to detect MEE and found particularly between 1-3 kHz that reflectance was higher in ears with suspected MEE compared to normal ears.

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 negative middle-ear pressure). The children were aged 5-7 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 than the criterion at 1.25 kHz (hit rate 100%), for a false-alarm rate of 10%. They also showed that reflectance was much more sensitive than 226 Hz tympanometry in detecting MEE. On average, ears with negative middle-ear pressure had higher reflectance 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, with decreased absorbance between 1.5 and 3 kHz in ears with surgically verified MEE. They also looked at admittance magnitude and phase. ROC analysis indicated that a slight improvement might be found by combining measures, but the key discriminative information is found in the absorbance function and there is little more to be found in the other measures because they are not statistically independent.
Summary (Nakajima, Rosowski, Shahnaz, & Voss, 2013) (Prieve, Feeney, Stenfelt, & Shahnaz, 2013) reviewed 8 studies that investigated WAI as a method to predict CHL (defined by ABG).

\H3\ Otosclerosis

Stapes fixation/otosclerosis tends to produce higher than average reflectance at lower frequencies, but may be within normal limits, or even lower than normal. Shahnaz et al. (2009) found increased reflectance below 1 kHz in otosclerotic ears, compared to a normal group. This is due to increased stiffness of the ME at the stapes, which increases the total stiffness of the ME measured at the eardrum. They found that reflectance 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 below 1 kHz. A combination of reflectance and tympanometry was able to detect all otosclerotic ears for the price of a higher false-alarm rate. Now although these finding show the ability to detect otosclerotic ears, Shahnaz et al used a group of ears with known otosclerosis. Of interest is whether otosclerosis can be differentiated from other middle-ear disorders, which is the situation clinicians’ face.

\H3\ Ossicular discontinuity

Ossicular discontinuity can present with a notch in the reflectance curve below 1 kHz. The disruption or near-disruption in an ossicle joint creates an extreme resonance, which
appears as a notch in the reflectance, or a peak in the absorbance. Although the absorbance is elevated, this power is *not* transmitted to the inner ear, but is dissipated in the resonant joint. We see this in some other pathologies, such as eardrum perforation, where power is dissipated in the ME cavity at the resonant frequency, not transmitted to the inner ear (like when you blow over the opening of a bottle and it sounds a tone.)

**Semicircular canal dehiscence (SCD)**

Ears with SCD tend to show a notched reflectance curve around 1 kHz, compared to normal ears (Nakajima et al., 2012), and which is higher in frequency than typically seen with ossicular discontinuity, and smaller and wider. SCD is often referred to as a “third window” (in addition to the oval and round windows of the cochlea), caused by a space where there is bone loss. This can be modelled as an additional cavity down the line in the transmission line model of the middle + inner ear, and it affects the local resonance that is often seen near 1 kHz. [Sarah – I haven’t found a description for the phys/phys reason why. I’m sure I’ve seen it but can’t find it.]

**Tympanic membrane perforations or PE tubes**

Figure 7. Four examples of TM perforation in adult ears, compared to the Rosowski norms.

Tympanic membrane perforations or pressure-equalization tubes tend to produce reflectance curves that are highly variable across frequency, with low reflectance at low frequencies. In
addition, the equivalent ear canal volume is typically >> 3cc. 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 a perforation.

As in the case of ossicular discontinuity, although the absorbance is elevated due to TM perforation, this power is *not* transmitted to the inner ear, but is dissipated in the ME cavity. The resonant frequency at which we see dissipation is related to the ME cavity size *and* the size of the hole (these two parameters can be modeled as a Helmholtz resonator, 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. Curiously, in a cadaveric ear, smaller perforations were more easily seen with WAI than larger ones (Nakajima et al., 2013; Voss, Merchant, & Horton, 2012), presenting as very low reflectance around 1 kHz.

### Negative middle-ear pressure and Eustachian tube dysfunction

NMEP can occur from ETD and otitis media and its presence stiffens the middle ear. Large degrees of ME pressure distinctly show in the reflectance curve as increased, flat reflectance, especially below 2 kHz. ME pressure cannot be directly estimated from WAI as is the case with tympanometry (i.e., from tympanic peak pressure readings). The degree of MEP is associated with degree of reflectance, but this is only noticeable when considering changes in a subject or perhaps in group averages – it is not easy to detect degree of NMEP from single measurements unless it is severe. It is important to evaluate presence of NMEP as it can affect other measurements, especially OAEs (even with “clinically normal” amounts of NMEP).

### Differential diagnosis of CHL in adults

It should now be apparent that although some pathologies are easy to detect from normal, they can be quite hard to tell apart from one another because their measurable effects are similar,
despite having different causes. For instance, otosclerosis and negative middle-ear pressure 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 on its own able to 

differentially diagnose all ME disorders, however in conjunction with other tests (such as 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). They 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 air-bone gap (ABG) audiogram to aid in the differential diagnosis in an office setting.

The absorbance level difference (ALD) used by Nakajima et al. (2012), is calculated by subtracting the absorbance (in decibels) from the mean normative absorbance value of -3.42 dB, derived in the companion paper by Rosowski et al. (2012), which considered normal rather than pathological ears. The ALD averaged over 0.6-1 kHz is the key WAI statistic. The ABG they used for the differential diagnosis is 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 air-bone gap (ABG) on pure-tone audiometry (averaged over either 500, 1000, 2000 Hz or 250, 500, 1000 Hz), and with an intact TM and aerated middle ear.

Nakajima et al. (2012) found:

- Ears with ALD (0.6-1kHz) < 1 dB and ABG (1-4kHz) > 10 dB were associated with stapes fixation.
- Ears with ABG (1-4kHz) <= 10 dB were associated with SCD.
Ears with ADL (0.6-1kHz) >= 1 dB and ABG (1-4kHz) > 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), but they are based on a small number of subjects (N=31 ears). These results suggest a larger study is warranted to further refine the differential diagnostic potential of WAI and audiometry.

**Sarah’s work**

**Wideband Tympanometry**

Our focus here has been on WAI made at ambient pressure. A variation involves pressurizing the ear canal as in tympanometry, the result being reflectance as a function of frequency and pressure, typically represented as a three dimensional magnitude plot. Advantages to this technique include the ability to slice the results by frequency or by pressure. Disadvantages include a much more complicated set of information to parse into clinical decision making framework, the need to pressurize the ear canal, and the effects pressurization can have on subsequent measurements (preconditioning) (Burdiek & Sun, 2014). Clinical efficacy is still being established; for example, Keefe, Sanford, Ellison, Fitzpatrick, and Gorga (2012) did not find an advantage to adding pressurization in detecting CHL in children.

**What the future holds…**

**Clinical advantages**

Rich field for theoretical and clinical research

**Potential topics**
/H1/ Acknowledgments

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