Parametric model of young infants’ eardrum and ear canal impedances supporting immittance measurement results. Part II: Prediction of eardrum and ear canal impedances for frequent pathological middle ear conditions

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Abstract – In order to gain a better understanding of wideband acoustic immittance (WAI) measurements, in this second part of a two-part paper, the parametric electro-acoustic model of the ear canal and the middle ear of young infants proposed in the first part is extended. The extension allows predictions of the influence of the pathological middle ear conditions middle ear effusion and negative static air pressure difference between the middle ear and the eardrum. Comparisons of the acoustic input impedance of the ear predicted by the model with real ear measurements in young infants’ ears with middle ear effusion show that the effects due to the pathology can be predicted well. For the negative static air pressure, a modeling approach was proposed but could not be confirmed yet, due to a lack of available measurement data. Furthermore, comparisons between different middle ear states (healthy, middle ear effusion and static air pressure difference) predicted by the model showed characteristic differences in all relevant WAI measures. However, it is also shown that WAI measures requiring an estimate of the cross-sectional area at the measurement position, i.e., absorbance and reflectance, are highly sensitive to this estimate.

Keywords: Acoustic impedance, Absorbance, Middle ear, Outer ear, Screening

1 Introduction

The aim of this work is to investigate the use of the model presented in [1], which predicts wideband acoustic immittance (WAI) measured within the ear canal, for detecting certain pathological middle ear conditions in young infants. WAI comprises measures like the acoustic impedance, the acoustic admittance, as well as measures derived from these quantities like the reflectance or the absorbance. WAI has been identified to be a promising tool in middle ear diagnostics [2]. In this context it is important to note that several properties of the ear canal and the middle differ between young infants on the one hand and older children and adults on the other hand, resulting in significantly different WAI results [1].

In the accompanying paper [1], a parametric electro-acoustic model of the ear canal and the middle ear of healthy young infants was proposed. The model parameters represent physiological properties wherever possible. The model was used to predict the ear canal acoustic impedance $Z_{ec}$, the quantity which is measured in a WAI measurement, and the acoustic input impedance at the eardrum $Z_D$, the interesting quantity in terms of middle ear diagnostics which can, however, not directly be measured. In the model, $Z_D$ is defined to be the acoustic input impedance of the middle ear including the eardrum. $Z_{ec}$, on the other hand, is defined to be the quotient of the sound pressure to the volume velocity at the measurement position in the ear canal. Three main findings were identified in the study.

- The soft and flexible ear canal walls in young infants’ ears affect $Z_{ec}$ up to about 1.5 kHz with no significant effect for higher frequencies.
- At medium frequencies around 1.8 kHz, $Z_{ec}$ is dominated by the input impedance at the eardrum $Z_D$. Thus, the largest effects of pathological middle ear conditions on $Z_{ec}$ is expected at medium frequencies.
- At high frequencies, the model prediction of $Z_{ec}$ has a much smaller magnitude compared to $Z_D$ indicating that $Z_{ec}$ is dominated by ear canal properties and therefore, it might be difficult to detect different middle ear conditions at these frequencies. However, the predictions must be taken with caution at high frequencies because many measurements on real ears...
showed higher $Z_{ac}$ magnitudes than the predictions. The relatively smaller difference between the measured $Z_{ac}$ magnitudes and $Z_0$ could mean that in some cases middle ear conditions could still be detected.

The objectives of the present paper are a) the extension of the model in order to predict the influence of different pathological middle ear conditions on the acoustic input impedances of the ear canal and the eardrum, and b) the investigation of implications for WAI measurements. Two pathological middle ear conditions have been chosen for the model extension:

- Firstly, middle ear effusion, resulting from otitis media, is the most frequent pathological middle ear condition in young infants with prevalence rates of 74% found in [3] for children in the first 6 months of life and 48.8% found in [4] for children aged between 2 and 6 months.
- Secondly, a negative static air pressure difference between middle ear and ear canal will be considered. It is often one of the earlier symptoms of otitis media (see e.g. [5, 6]).

The model predictions will be compared with ear canal acoustic impedances measured in infant ears published in [7]. The aspects of the measurement procedure relevant for this study will be described here in detail. As described in [1], the measurement method is based on [8], extended by consideration of discontinuity and end corrections in [9]. The method including the calibration procedure is described in detail in [9]. The measurements were performed using a custom-made impedance probe to measure the ear canal impedance on infant ears. The probe was designed to avoid over-pressure in the ear canal caused by inserting it into the ear canal at the cost of a decreased signal-to-noise ratio (SNR) at frequencies below about 1 kHz, further details can be found in [7]. This was realized by a pressure equalizing duct with an inner diameter of 0.6 mm.

The subjects who participated in the study were aged between 2 weeks and 5 months. It should be noted that the infant ear undergoes developmental changes within the first 6 months of life and 48.8% found in [4] for children aged between 2 and 6 months.

Fluid in the middle ear may occur as remaining amniotic fluid after birth or as middle ear effusion. In both cases, a substantial part of the middle ear cavities is filled with liquid, covering partly the eardrum.

The representation in the model should comprise a reduced volume of the tympanic cavity, resulting in a smaller value of the acoustic compliance of the tympanic cavity $N_{cav}$, and a reduced area of the eardrum that is vibrating, resulting in modified values of the effective eardrum area $A_0$, the eardrum acoustic shunt impedance $Z_{ac}$, and the mechanical eardrum parameters, namely the mechanical mass and resistance of the free vibrating portion of the eardrum $m_{free}$ and $w_{free}$.

A first approach to implement the effect of fluid in the middle ear is proposed as follows: The effective eardrum area is decreased using a factor $x_A < 1$, such that

$$A_0|_{fluid} = x_A A_0.$$  \hspace{1cm} (1)

According to [1] (Eq. 21)

$$Z_{ac} = W_{ac} + j0M_{ac} + (j0N_{ac})^{-1},$$  \hspace{1cm} (2)

the eardrum acoustic shunt impedance $Z_{ac}$ is determined by the acoustic compliance $N_{ac}$, the acoustic mass $M_{ac}$, and the acoustic resistance $W_{ac}$. For simplicity it is assumed that the eardrum geometry remains circular when the area is decreased, i.e., only the radius of the vibrating membrane is reduced by $\sqrt{x_A}$. From [1] (Eq. 22),

$$N_{ac} = \frac{\pi}{8} \frac{a_{ed}^3}{T_{0,ed} h_{ed}},$$  \hspace{1cm} (3)

with $a_{ed}$ the radius of the eardrum, $T_{0,ed}$ the eardrum tension and $h_{ed}$ the thickness of the eardrum, one can infer that the acoustic compliance should change with $x_A^2$, i.e., the acoustic compliance decreases if the eardrum area decreases. The acoustic mass should then change with $1/x_A$ according to [1] (Eq. 23),

$$M_{ac} = \frac{4}{3} \frac{\rho_{ed} h_{ed}}{\pi a_{ed}^3},$$  \hspace{1cm} (4)
with \( \rho_{\text{ed}} \) the membrane mass density, i.e. the acoustic mass increases if the eardrum area decreases. Because a simple relation between the acoustical resistance and the area of the eardrum is not known, in a first approach \( W_{\text{ac}} \) is kept constant. The mechanical mass of the free vibrating part of the eardrum \( m_{\text{free}} \) is changed by multiplying it with \( x_A \) assuming that the free vibrating portion of the eardrum decreases with \( x_A \). A relation between the eardrum area and the mechanical resistance \( w_{\text{free}} \) is not known, therefore the value is kept constant.

The volume change of the tympanic cavity can also be linked to the parameter \( x_A \), by letting

\[
V_{\text{tav}}|_{\text{fluid}} = V_{\text{tav}} \cdot x_A, \tag{5}
\]

effectively assuming that the volume of the tympanic cavity results from protruding the reduced area of the vibrating eardrum.

The impact of different values of \( x_A \) on the acoustic input impedance at the eardrum \( Z_D \) and the eardrum acoustic shunt impedance \( Z_{\text{ac}} \) is depicted on the left side of Figure 1. It can be seen that the smaller the eardrum area gets, the more \( Z_D \) is dominated by \( Z_{\text{ac}} \). Furthermore, a smaller eardrum area shifts the magnitude minimum and the phase change to higher frequencies. At frequencies smaller and larger than the minimum, the magnitude increases with decreasing eardrum area.

In the middle column of Figure 1, the ear canal impedance \( Z_c \) is depicted for different values of \( x_A \). The ear canal is modeled using a constant ear canal radius of 1.7 mm and an ear canal length of 14 mm. In the right column of Figure 1, measurement results from \([7]\) of ear canal impedances measured in infants’ ears with diagnosed middle ear effusion are depicted. The measurements suffer from a bad signal-to-noise ratio (SNR) at low frequencies, therefore, only those values are depicted in which the coherence between the signal applied to the probe speaker and the signal sensed by the probe microphone exceeded 0.5. Another effect that is present in the majority of the measurements is acoustic leakage resulting in an impedance magnitude increasing with increasing frequency and positive phase values up to about 2 kHz. In the modeling, a tightly sealed ear canal is assumed, i.e. without leakage, therefore, in the following comparisons between measurements and predictions concentrate on higher frequencies, while differences up to about 2 kHz are ignored. The predicted shift of the first magnitude minimum below 2 kHz to higher frequencies is well reflected in the measurements. Above this first minimum, there is a mismatch between model and measurements. In the model output, the second

**Figure 1.** First approach to model the effects of fluid in the middle ear. Left: Acoustic input impedance at the eardrum \( Z_D \) and eardrum acoustic shunt impedance \( Z_{\text{ac}} \) for different factors of eardrum area. Middle: Ear canal impedance for different factors of eardrum area resulting from the model using an ear canal radius of 1.7 mm and an ear canal length of 14 mm. Right: Ear canal impedances measured in infants’ ears with middle ear effusion from \([7]\).
minimum, which is mainly determined by the ear canal geometry, is only weakly affected by the variation of $x_A$. In the measurements, however, no second minimum can be seen up to 10 kHz. The only magnitude minimum in the measurements is broader than the first minimum of the model. It is located at lower frequencies than the second minimum of the model.

Based on these observations, the previously described approach of modeling the effects of fluid in the middle ear is amended. Firstly, in order to obtain a greater shift of the magnitude minimum of $Z_{ac}$, the acoustic mass $M_{ac}$ is not changed with changing $x_A$, which can be interpreted as the acoustically effective part of the vibrating mechanical mass of the eardrum being proportional to $x_A^2$ instead of $x_A$, which was expected in the first place. Secondly, the acoustic resistance $W_{ac}$ is decreased with decreasing $x_A$, using

$$W_{ac}|_{\text{fluid}} = x_A W_{ac}. \quad (6)$$

The result of this final approach used to model the effects of fluid in the middle ear can be seen in Figure 2. Again, on the left side the acoustic input impedance at the eardrum $Z_D$ and the eardrum acoustic shunt impedance $Z_{ac}$ are depicted for different factors of eardrum area. Middle: Ear canal impedance for different factors of eardrum area resulting from the model using an ear canal radius of 1.7 mm and an ear canal length of 14 mm. Right: Ear canal impedances measured in infants’ ears with middle ear effusion from [7].

3 Predicting the effect of negative pressure difference between the middle ear cavities and the ear canal

Due to a dysfunction of the eustachian tube and the gas resorption of the mucosa, the static pressure of air in the
middle ear cavities $p_{0,ME}$ can be decreased. Compared to the ambient static pressure of air $p_0$, the pressure difference $\Delta p_0 = p_{0,ME} - p_0$ gets negative. Consequences of the negative $\Delta p_0$ are a retracted eardrum with an increased tension of the membrane, a reduced density of air in the middle ear cavities, and a slightly decreased volume of the tympanic cavity. A relevant range of $\Delta p_0$ can be obtained from the measurement range in tympanometry, which reaches down to $-400$ daPa ($-4$ kPa).

In the model, the tension of the eardrum is represented in the acoustic compliance $N_{ac}$, which is part of the eardrum acoustic shunt impedance $Z_{ac}$. As described in part I [1], Sect. 2.2.3, $Z_{ac}$ is modeled as a stretched circular membrane [1]. According to [14] (p. 68), the displacement $\xi_z$ orthogonal to the membrane-plane due to a pressure difference $\Delta p_0$ at a normalized radial distance from the center $r/a$ is given by

$$\xi_z(r/a) = \Delta p_0 \frac{a^2}{4T_0h} \left(1 - \left(\frac{r}{a}\right)^2\right),$$

with $T_0$ the tension of the non-displaced membrane and $h$ the membrane thickness. Therefore, the displacement at the center of the membrane is given by

$$\xi_z(0) = \Delta p_0 \frac{a^2}{4T_0h}.$$  

(8)

The membrane takes the shape of an elliptical paraboloid. The area of the membrane with the displacement $\xi_z(0)$ is given by

$$A' = \pi a^2 \frac{a}{6\xi_z(0)} \left((a^2 + 4\xi_z^2(0))^2 - a^4\right).$$

(9)

corresponding to an area of a circle with a radius of

$$a' = \sqrt[3]{\frac{a}{6\xi_z(0)} \left((a^2 + 4\xi_z^2(0))^2 - a^4\right)}.$$  

(10)

The in-membrane strain $S$ for a given displacement $\xi_z(0)$ is given by

$$S = \frac{a'}{a} - 1,$$

(11)

resulting, by using equations (8) and (10), in

$$S(\Delta p_0) = \sqrt{\frac{(a^2 + 4\left(\Delta p_0 \frac{a^2}{4T_0h}\right)^2)^2}{6a\left(\Delta p_0 \frac{a^2}{4T_0h}\right)^2} - 1}.$$  

(12)

With the relation between strain and tension of the membrane given in [14]

$$T = \frac{E}{(1 - v)} S,$$

(13)

with Young’s modulus $E$ and Poisson’s ratio $v$, the membrane tension $T_0$ for a static $\Delta p_0$ can be calculated using

$$T_0 = T_0 + \frac{E}{(1 - v)} \sqrt{\frac{(a^2 + 4\left(\Delta p_0 \frac{a^2}{4T_0h}\right)^2)^2}{6a\left(\Delta p_0 \frac{a^2}{4T_0h}\right)^2} - 1}.$$  

(14)

The acoustic compliance of the displaced membrane is then given by

$$N'_{ac} = \frac{\pi}{8} \frac{\rho_{el} a_{el}^4}{T_{0,el} h_{el}},$$

(15)

with the thickness $h_{el}$ of the stretched eardrum. If an incompressible isotropic medium of the membrane ($v = 0.5$, see e.g. [15]) is assumed, the volume of the membrane remains unchanged. With equation (11), the thickness of the stretched eardrum is given by

$$h_{el}' = \frac{h_{el}}{(1 + S)^2}.$$  

(16)

As mentioned above, the negative pressure difference affects the middle ear cavities by decreasing the density of air in...
the middle ear, and by decreasing the volume of the tympanic cavity. The decrease in density of air in the middle ear cavities increases the compliance of the tympanic cavity \( N_{\text{tav}} \) and the compliance of the antrum \( N_{\text{an}} \). The density is given by

\[
\rho_{\text{tav}} = \rho_{\text{an}} \left( \frac{p_0 + \Delta p_0}{p_0} \right). \tag{17}
\]

The volume occupied by the statically displaced membrane, i.e. the volume decrease of the tympanic cavity, is given by

\[
\Delta V_{\text{tav}} = \frac{\pi}{2} r_{\text{el}}^2 \Phi(0). \tag{18}
\]

Both effects, the decrease of \( \rho_{\text{tav}} \) and the decrease of \( V_{\text{tav}} \) are very small. Especially the resulting change of the cavity volume and its density decreases the numerator whereas the density change decreases the denominator.

The only parameter value needed to model a negative pressure difference between the middle ear cavities and the ear canal, is the Young’s modulus \( E \) of the membrane (see Eq. (15)). Note that Poisson’s ratio \( \nu \) was already chosen to be 0.5. Unfortunately, there is no easy way to determine \( E \). In the literature, values of Young’s modulus of real tympanic membranes can be found, at least for adult ears. However, those values cannot be used directly for the modeling approach of a stretched circular membrane because of the different membrane shapes and in-plane tensions. This means that a value of \( E \) could only be determined indirectly, e.g. from impedance measurements at different strain levels. It would, however, be necessary that the effect of membrane stress could be separated from other effects like e.g. those caused by compliant ear canal walls in tympanometry. To the best of our knowledge, no studies exist in which the effect of negative pressure difference in young infants’ middle ears on immittance measurements is investigated independently of the influence of the soft and flexible ear canal walls. Therefore, we had to estimate a reasonable value of \( E \). Assuming that \( E \) has a substantial effect on \( Z_0 \) for \( \Delta p_0 \) in the range from 0 to \(-2.5 \) kPa, we chose a value of \( E = 100 \) MPa. On the left side of Figure 4, both, the eardrum acoustic shunt impedance \( Z_{\text{sc}} \), and the acoustic input impedance at the eardrum \( Z_{\text{D}} \) are depicted for different \( \Delta p_0 \). As can be seen, the minimum in the magnitude of \( Z_{\text{sc}} \) shifts to higher frequencies when \( \Delta p_0 \) is decreased from zero to negative values. This can also be observed for the global minimum in \( Z_0 \). If the minimum in \( Z_{\text{sc}} \) is shifted to higher frequencies, a new magnitude minimum in \( Z_0 \) at about 800 Hz emerges followed by a maximum at about 1.2 kHz. This minimum is caused by the mechanical middle ear components, in particular by the ossicles. The ear canal impedance \( Z_{\text{ec}} \) of the model is depicted on the right side of Figure 4. Additionally, measurements from [7] on infants’ ears where the tympanometric peak pressure was smaller than \(-1 \) kPa (\(-100 \) daPa) are depicted, unfortunately, there where only two ears in which this was clearly the case. As can be seen in the magnitude, a local maximum appears at about 1.1 kHz if \( \Delta p_0 \) gets negative. This can also be observed in the measurements. In the phase of \( Z_{\text{ec}} \), the increase with frequency between 1.2 kHz and 2.5 kHz is shifted to higher frequencies with a steeper slope for negative \( \Delta p_0 \). Regarding the modeling approach, an assessment of the choice of parameters is not possible at this time due to a lack of available measurement data, therefore, it can only be seen as a first proposal. However, the resulting effects show at least some similarities with the few measurements.

In summary, an approach to model the effect of negative pressure difference between the middle ear cavities and the ear canal was proposed. However, there are several simplifications in the modeling approach. First of all, using a stretched circular membrane is obviously a significant simplification, which among other things has the consequence that a physiologically correct value of Young’s modulus cannot be used. Furthermore, the assumption of the eardrum behaving linear-elastically may not fully hold in practice, even in the range of typically observed tympanometric pressures. Despite these assumptions some characteristics observed in the few available measurements are captured by the model. More data involving negative tympanic pressures would be required for a more in-depth validation.

4 Comparison of immittance measures for normal and pathological middle-ear conditions

There is an on-going discussion of the suitability of different immittance measures for screening or diagnostic purposes of young infants’ middle ears. All these measures are derived from the acoustic impedance in the ear canal \( Z_{\text{ec}} \) or its reciprocal, the acoustic admittance. Most investigations concentrate either on the energy reflectance (also called power reflectance) which is given by \( |R|^2 = \frac{|(Z_{\text{ec}} - Z_{\text{w}})/(Z_{\text{ec}} + Z_{\text{w}})|^2}{1 - |R|^2} \), with the tube wave impedance \( Z_{\text{w}} \) of the ear canal cross-section at the measurement position, or on the energy absorbance given by \( 1 - |R|^2 \).

Sometimes, the phase or group delay of the pressure reflectance \( R \) is also discussed [16, 17].

The model output for different immittance measures is depicted in Figure 5 for three different states of the middle ear, namely normal, fluid in the middle ear \( (x_{\lambda} = 0.2) \), and a negative pressure difference of \( \Delta p = -2500 \) Pa between the middle ear cavities and the ear canal. In the left column, the acoustic input impedance at the eardrum \( Z_{\text{D}} \) is depicted. \( Z_{\text{D}} \) directly reflects the state of the middle ear and hence is the desired measure, however, it is not directly available. In the second column of Figure 5, the acoustic input impedance \( Z_{\text{ec}} \) is depicted as it would be measured in the ear canal. The complex-valued pressure reflectance \( R \) is depicted in the third column with its magnitude (top) and its group delay (bottom). Finally the energy reflectance is depicted in the top and the energy absorbance in the bottom panel of the last column. All of the selected
Impedance measures show characteristic differences between the three states.

The most important changes compared to the normal middle ear condition in the case of fluid in the middle ear are a massively increased magnitude of $Z_D$ up to a frequency of 6 kHz and a large difference in the phase of $Z_D$ between 1.5 kHz and 8 kHz. The magnitude of $Z_{ec}$ differs largely between 500 Hz and 2 kHz where a local maximum appears and at high frequencies due to a shift of the minimum to lower frequencies. The phase of $Z_{ec}$ has a significantly different slope at frequencies higher than 500 Hz.

In the case of a negative pressure difference between middle ear and ear canal, the magnitude of $Z_D$ is significantly larger between 900 Hz and 2 kHz. The phase of $Z_D$ has a local maximum at 900 Hz and the group delay has significantly smaller values between about 1 kHz and 3 kHz.

In conclusion, all of these quantities should be included in the search for an objective classification of middle-ear disorders in humans. This is particularly true for phase (or alternatively, group delay) quantities, which have mostly been ignored so far.

Measurement-based data of the mean energy absor-}


cance for age groups between 1 and 3 months have been published in [12, 18, 19, 17]. In Figure 6 these data are depicted with thin lines together with the energy absor-}

cance predicted by the model (thick lines) using three different areas to compute the tube wave impedance: 1) the correct ear canal cross-section with a diameter of 3.4 mm, 2) an area with a diameter of 4.5 mm, and 3) an area with a diameter of 7.5 mm. The selection of diameters is based on values used in the commercially available devices for WAI measurements.

Some agreements can be found in all predictions and measurement-based data: At low frequencies absorbance increases with frequency, and a maximum value at about 2 kHz is followed by a notch between 3 Hz and 4 kHz. However, large differences can be seen in both, predictions and measurement-based data. As can be seen in the predictions, the reference area used for the tube wave impedance largely affects the value of the energy absorbance in the whole
frequency range between 100 Hz and 10 kHz. It should be noted that these differences in energy absorbance are only the result for one single idealized (cylindrical) ear canal geometry. They do not specify something like a normative range since other ear canal geometries will result in other differences caused by the selected reference area. In [12, 17] an acoustic estimate of the ear canal area at the probe tip was used, and in [18, 19] a reference area with a diameter of 7.5 mm was used. It can be seen that the measurement-based data differ and that effects of age (e.g. the effect of compliant ear canal walls) can possibly not be distinguished from effects due to a mismatch in reference area. The model predictions suggest that, if the energy absorbance is used as the immittance measure of choice, it might be more valuable to look at the variation with frequency in a range from 1 to 6 kHz rather than to look at absolute values.

5 Discussion

The parametric electro-acoustic model of young infants’ eardrum and ear canal impedances proposed in [1] has been extended to account for selected pathological middle ear conditions. Firstly, the condition of fluid in the middle ear, which is the most frequent pathological middle ear condition in young infants, has been considered in the model. It has been realized by a single factor $x_A$ decreasing the effective eardrum area and the volume of the tympanic cavity. Secondly, the condition of negative pressure difference $\Delta p$ between the middle ear cavities and the ear canal has been implemented. Comparisons between model predictions for a normal middle ear status and for these pathological middle ear conditions showed characteristic differences for all WAI measures, including quantities that have mostly been ignored, i.e. phase quantities. Furthermore, it was shown that different reference diameters used to compute the acoustic absorbance lead to large differences for the same ear.

5.1 Implications for WAI measurements

For those immittance measures which can easily be determined by acoustic measurements in the ear canal ($Z_{ec}$ and $R$), the effects of the different pathological middle ear conditions showed large differences at frequencies
between 600 Hz and 6 kHz. In part I \[1\] it was found that the ear canal properties (geometry and compliant walls) dominate $Z_{ec}$ at frequencies below 1.5 kHz and above 4 kHz, therefore, the frequency range from 1.5 kHz to 4 kHz should be suited to detect pathological middle ear conditions in young infants’ ears. This is in agreement with results of studies in which WAI was measured on young infants’ ears classified to have either a normal or a pathological middle ear status \[7, 18, 20\]. In \[18\], the energy reflectance ($|R|^2$) measured on infants aged between 0.7 and 5.9 months (median 2.1 months) was compared. It was found that a discrimination between normal hearing and conductive hearing loss was possible at frequencies around 1.6 kHz. In \[20\], the absorbance ($1 - |R|^2$) was measured on infants aged from birth up to 4 months (mean 1.3 months). A mixed-model analysis of variance (ANOVA) showed significant different mean absorbance values between the normal hearing group and combined groups of ears having a conductive or a mixed hearing loss at frequencies from 1 to 8 kHz. In \[7\], the ear canal impedance ($Z_{ec}$) was measured on infants aged between 0.5 and 4.8 months (mean 1.8 months). Characteristic differences of $Z_{ec}$ between normal ears and ears with middle ear effusion were found in the frequency range from 1 to 5 kHz. All these studies show that discriminating between normal and pathological middle ears of young infants based on WAI seems to be possible. Therefore, the goal should be the specification of normative WAI data for young infants. However, as was shown in Section 4, reflectance measures are strongly sensitive to the choice of the reference area. In order to make WAI measures less dependent on this choice (and thus less dependent on the measurement device being used), impedance (or admittance) quantities which do not require area information should be used to arrive at normative data. When reflectance quantities are used in studies, it is essential that complex quantities are specified and that the assumed reference area is also specified to at least allow comparisons with other data.

5.2 Age dependency

The model initially proposed in \[1\] and extended in this paper targets young infants aged between 1 and 5 months. The parameter values are partly based on physiological data found in literature. However, the few physiological data available do not allow a more detailed age dependency to be implemented yet. Comparisons of model predictions have been made with ear canal impedances measured on infants aged between 0.5 and 4.8 months. Our model does not account for age dependent effects within these age range yet.

Some age-dependent WAI data can be found in the literature. In \[12\] absorbance values for the age groups of 1, 3, and 6 months showed that up to about 630 Hz the absorbance significantly decreases with increasing age. This was explained by a decrease of the effects due to the soft and flexible ear canal walls. In \[19\], absorbance values for the age groups roughly at 0.5, 2, 3, 4, 5 and 6 months showed that below about 500 Hz the absorbance slightly decreases with increasing age. At high frequencies around 3–4 kHz, the absorbance increases with increasing age. A comparison of absorbance values for age groups of 1 month and 6 months shows that an absorbance decreasing with increasing age at low frequencies was found in \[12, 17, 19\]. However, in \[12\] this low-frequency-behavior was observed up to about 2 kHz, in \[19\] this was only true up to 500 Hz. In \[17\], in turn, the decreasing absorbance with increasing age was observed in the whole frequency range, but in \[12, 19\] an absorbance increasing with increasing age was observed at frequencies around 3–4 kHz. It can be concluded that: 1) effects caused by the soft and flexible ear canal walls decrease in the first months of life. 2) age

![Figure 6](image-url)

**Figure 6.** Energy absorbance predicted by the model using different areas to compute $Z_{tw}$, together with measurement based data from \[12, 18, 19, 17\] for different age groups with the mean or median age in days given in parentheses.
dependency at high frequencies around 3–4 kHz can be expected. 3) unfortunately, the results of various studies in which absorbance values were determined for age groups contradict each other.

Furthermore, in [17] median absorbance values for newborns and for infants aged 1, 6, 9, and 12 months were published. Significant differences were found among the group of newborns, the age group of 1 month and the group 6–12 months. For clinical applications, the authors recommended to use separate normative references for these age groups. Therefore, in the future, an extension of our model to newborns might be valuable for the application of WAI in combination with the universal newborn hearing screening (UNHS). In this context, the effects of fluid in the middle ear are of special interest because UNHS-tests can be affected if amniotic fluid in the middle ear are of special interest because UNHS-tests can be affected if amniotic fluid is still present in the middle ear shortly after birth.

6 Concluding remarks

In this paper, the parametric electro-acoustic model of eardrum and ear canal impedances of young infants [1] was extended to account for specific pathological middle ear conditions:

- The effect of fluid in the middle ear on \( Z_{ec} \) was modeled by a single factor \( f \) decreasing the effective eardrum area and the volume of the tympanic cavity. A factor of about 1/5 showed a good estimation of the effects of fluid in the middle ear observed in measurements on infants’ ears aged between 0.5 and 4.8 months.
- For the pathological middle ear condition of negative pressure difference \( \Delta p_0 \) between the middle ear cavities and the ear canal, a modeling approach was proposed. Characteristic effects in the modeled \( Z_{ec} \) caused by a negative \( \Delta p_0 \) were found in the medium frequency range. At about 1.1 kHz, a local magnitude maximum appeared if \( \Delta p_0 \) got negative. The increase of the phase value between 1.2 kHz and 2.5 kHz was shifted to higher frequencies. A confirmation of these effects by measurements was not yet possible due to a lack of available measurement data involving negative tympanic pressures. This could be part of future work.

Comparisons between the middle ear states normal, fluid in the middle ear and static pressure difference between ear canal and middle ear showed characteristic differences in all relevant immittance measures predicted with the model. Furthermore, it was shown that different reference diameters which are used in different measurement devices to compute the absorbance or energy reflectance lead to large differences in these immittance measures. Hence, the results are not comparable and should not be used to define normative data. Instead, quantities that do not require area information should be used, such as the acoustic impedance.

In the future, age-dependent adaptations of the model could be valuable, especially to cover newborns. A further pathological condition to be included might be an oedematous middle ear. Furthermore, a middle ear classifier based on impedance (both magnitude and phase) will be developed.

Conflict of interest

The authors declare no conflict of interest.

Data availability statement

An implementation of the complete model is available on GitHub, under the reference https://github.com/tobiassankowsky/acoustic_impedance_infant_ear.

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