Predictions of middle-ear and passive cochlear mechanics using a finite element model of the pediatric ear

Xuelin Wang,¹ Douglas H. Keefe,² and Rong Z. Gan^{1,a)}

¹*School of Aerospace and Mechanical Engineering and Biomedical Engineering Center, University of Oklahoma, Norman, Oklahoma 73019, USA*

²*Boys Town National Research Hospital, Omaha, Nebraska 68131, USA*

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A finite element (FE) model was developed based on histological sections of a temporal bone of a 4-year-old child to simulate middle-ear and cochlear function in ears with normal hearing and otitis media. This pediatric model of the normal ear, consisting of an ear canal, middle ear, and spiral cochlea, was first validated with published energy absorbance (EA) measurements in young children with normal ears. The model was used to simulate EA in an ear with middle-ear effusion, whose results were compared to clinical EA measurements. The spiral cochlea component of the model was constructed under the assumption that the mechanics were passive. The FE model predicted middle-ear transfer functions between the ear canal and cochlea. Effects of ear structure and mechanical properties of soft tissues were compared in model predictions for the pediatric and adult ears. EA responses are predicted to differ between adult and pediatric ears due to differences in the stiffness and damping of soft tissues within the ear, and any residual geometrical differences between the adult ear and pediatric ear at age 4 years. The results have significance for predicting effects of otitis media in children. © 2016 Acoustical Society of America.

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I. INTRODUCTION

Transmission of sound from the ear canal to cochlea is mediated by the mechanical properties of the ossicles, tympanic membrane (TM), middle-ear ligaments, middle-ear muscle tendons, and cochlear structures. Growth of the external and middle ear from a young child to adult increases the volumes of the ear canal and middle-ear cavity, and changes the orientations between ossicles (particularly in newborn infants) (Abdala and Keefe, 2012; Qi *et al.*, 2008). The developmental alterations of the human ear affect the dynamic characteristics of the peripheral auditory system and sound energy transmission into the cochlea (Eggermont and Moore, 2012). Understanding the acoustic response properties of the pediatric ear has important implications for interpretation of middle-ear assessment measures of pediatric hearing.

Biologically realistic finite element (FE) models of the human ear have been developed to understand the middle ear and cochlea functions in normal and pathological conditions in the past two decades (Gan *et al.*, 2004; Wang *et al.*, 2007; Zhao *et al.*, 2009; Zhang and Gan, 2011; Wang *et al.*, 2014). FE models were successfully employed to simulate various middle-ear disorders in adult ears such as otitis media (Gan and Wang, 2007), TM perforation (Gan *et al.*, 2009), and ossicular chain fixation or disarticulation (Huber *et al.*, 2003; Zhang and Gan, 2013). However, these existing human FE models are all based on the physical structure of adult ears, and may not accurately describe the structural

features of infant and young children ears. An exception is Qi *et al.* (2008), who presented a FE model of a 22-day-old newborn middle ear. The model was used to predict the volume displacement of the TM under static pressure variations, and the equivalent volume of the ear canal as a function of tympanometric frequencies between 0.2 and 1.0 kHz.

Otitis media is the most common inflammatory or infectious disease in young children (Leibovitz *et al.*, 2010). A FE model of the pediatric ear would be helpful to better understand effects of otitis media, such as middle-ear effusion (MEE), on the auditory system in young children. Adult FE models have been validated using temporal-bone measurements of umbo and stapes footplate motion or of middle-ear pressure transfer functions. However, the development of a FE model for a pediatric ear is constrained due to a comparative lack of sources for pediatric temporal bones and an absence of published measurements of soft tissue mechanical properties of the pediatric ear. Given these difficulties, an alternative validation process was developed for a pediatric FE model using acoustical ear-canal measurements obtained in human subjects.

Wideband energy absorbance (EA) or energy reflectance (ER) are non-invasive measures of the mechano-acoustic response of the middle ear to sound stimulation in the ear canal. These measures are used in research and clinical audiology with both adults and young children to differentiate between normal ears and ears with specific middle-ear disorders (Keefe *et al.*, 1993; Feeney *et al.*, 2003; Allen *et al.*, 2005; Beers *et al.*, 2010; Ellison *et al.*, 2012; Hunter *et al.*, 2013). EA is the fraction of incident energy absorbed by the ear at a probe inserted in a mid-canal location. To the extent that negligible sound

^{a)}Electronic mail: rgan@ou.edu

energy is absorbed along the walls of the ear canal, EA is equal to the fraction of incident energy absorbed by the middle ear at the TM. The EA was calculated for the adult ear by [Zhang and Gan \(2013\)](#) using an adult FE model for an ear with normal function, with varying middle-ear pressure, with varying amounts of MEE, and for such middle-ear disorders as a stiffened ossicular chain, ossicular discontinuity and a fixed-stapes condition.

The present study reports middle-ear transfer functions including EA, which were predicted by a pediatric FE model for ears with normal function and with MEE. This model included an ear canal, middle ear, and spiral cochlea based on histological section images of a pediatric temporal bone at age of 4 years. The middle-ear model parameter values were selected so as to accurately predict EA in comparison to published data in young children. The cochlear model parameters were selected so as to accurately predict transfer functions for basilar membrane (BM) displacement in adult human temporal bones under the assumption that cochlear function was mature by age 4 years. The model predictions of cochlear mechanics were generated by a model with passive, linear mechanics. That is, no active nonlinear properties of mechanics were included in the model. The results of the pediatric FE model were compared with predictions from published adult FE models. The model provided estimates of the changes in EA, TM motion, and BM displacement produced by a pediatric ear with MEE relative to the corresponding responses in a pediatric ear with normal function.

II. HUMAN SUBJECTS

A clinical database of published EA measurements in young children was analyzed in the present study with respect to model predictions. EA was measured in two “normal” groups of children who either had normal hearing and absence of otitis media history ([Keefe et al., 2012](#)) or normal middle-ear function ([Ellison et al., 2012](#)). The age range was 3 to 8 years with a mean age of 5 years in [Keefe et al.](#), and 0.5 to 7 years with a median age of 1.3 years in [Ellison et al.](#) Each of these age ranges included the age of 4 years corresponding to the age at which the temporal bone had been harvested.

The normal group in the [Keefe et al. \(2012\)](#) study included 43 ears from 26 subjects that satisfied the following criteria: no otitis media within the last six months, no history of surgery to insert pressure-equalization tubes in the TM, air-conduction thresholds not exceeding 15 dB hearing level (HL) at any octave frequency from 0.25 to 8.0 kHz, and air-bone gaps not exceeding 15 dB at any octave frequency from 0.25 to 4.0 kHz. The normal group in the [Ellison et al. \(2012\)](#) study included 59 ears of 44 children with normal pneumatic otoscopic findings, and no history of ear disease or middle-ear surgery.

EA was also measured in two “otitis media” or “MEE” groups of children. In [Keefe et al. \(2012\)](#), one such group included 35 ears of 24 patients with a conductive hearing loss who were medically classified as having otitis media with effusion. In [Ellison et al. \(2012\)](#), the other such group included 53 ears of 44 patients who were scheduled for myringotomy and tube placement, such that the EA test was

performed within one hour of their surgery. Data were included in this group only if the physician conducting the surgery confirmed the presence of MEE at operation. Each otitis media group was age-matched with its normal group.

III. FINITE ELEMENT MODEL

A. Anatomical reconstruction

A three-dimensional (3D) geometric model of the pediatric ear was created based on histological sections of a temporal bone (male, 4-year old, left) provided by the National Temporal Bone Lab at Massachusetts Ear and Eye Infirmary. The histological sections were scanned into images at 2400 dpi resolution for reconstruction of the geometry. All images were segmented to construct the 3D geometry of the ear model following the process of [Gan and Wang \(2007\)](#). Briefly, the ear-canal wall, middle-ear structures, and bony wall of the cochlea were identified. These included, in part, the TM, three ossicles (malleus, incus, and stapes), the incudomalleolar and incudostapedial (IS) joints, and the cochlear scalae [scala vestibula (SV), scala media (SM), and scala tympani (ST)]. The contour lines of these structures were marked by a series of key-points and their coordinates were extracted in SOLIDWORKS (SolidWorks, Inc., Concord, MA). The points were employed for constructing spline curves, and the surface skinning and smoothing techniques were then applied to generate closed surfaces and correct any distortion inherent in the slice images ([Sun et al., 2002](#)). Finally, the key-points and surface representations were inputs to the FE software ANSYS (ANSYS, Inc., Canonsburg, PA), which constructed the 3D solid geometry of the ear.

Table I lists the characteristic dimensions of the ear canal, middle-ear components, and middle-ear cavity measured from the model and compared with two adult ear models ([Gan et al., 2004](#); [Gan and Wang, 2007](#)). Similar to the approach used in constructing these adult models, the middle-ear cavity included only the tympanic cavity, and the mastoid air cells were not taken into account. The absence of mastoid air cells in the model was due to their complex morphology and the absence of detailed experimental data on their acoustical and mechanical properties. Although an equivalent mastoid cavity model has been used in FE analysis ([Lee et al., 2010](#)), the model reflected a smaller effect of mastoid air cells than that reported in experimental measurement ([McElveen et al., 1982](#)), and the model lacked an accurate morphometric description.

Table II lists the dimensions of the cochlear components. The cochlear duct was divided into the SV, SM, and ST by the basilar membrane and Reissner’s membrane (RM) and the length of BM or RM was 34.3 mm. Because the width and thickness of the BM and RM were not reliably imaged in the histological sections and distinguished from other soft tissue, the BM and RM were modeled using a rectangular cross-sectional geometry with the same dimensions as in a previous adult model ([Zhang and Gan, 2013](#)). The width of the BM varied from 100 μm at the base to 500 μm at the apex, and the BM thickness varied from 7.5 μm at the base to 2.5 μm at the apex. The thickness of RM varied from 6.6 μm at the base to 4.4 μm at the apex.

TABLE I. Dimensions and physical properties of the pediatric ear model (ear canal and middle ear) in comparison with published FE models of two adult ears.

Structure	Pediatric model	Adult model (Gan <i>et al.</i> , 2004)	Adult model (Gan and Wang, 2007)
Ear canal			
Length (mm)	25.00	30.22	29.65
Volume (mm ³)	1037	1657	1540
Angle between TM and meatus (deg.)	70.57	78.28	71.30
Tympanic membrane			
Diameter along manubrium (mm)	9.20	10.86	9.76
Diameter perpendicular to manubrium (mm)	8.84	9.24	9.01
Height of the cone (mm)	1.38	1.46	2.42
Surface area (mm ²)	66.45	72.01	83.73
Thickness (μm)	50–60	50–100	50–100
Malleus			
Length from manubrium end to lateral process end (mm)	4.00	4.71	4.76
Total length (mm)	7.20	8.11	9.46
Mass (mg)	28.11	30.42	30.01
Incus			
Length along long process (mm)	5.80	6.02	6.66
Length along short process (mm)	4.74	4.58	4.58
Mass (mg)	25.74	26.47	35.17
Stapes			
Height (mm)	2.85	2.66	4.18
Length of footplate (mm)	2.24	2.64	2.83
Width of footplate (mm)	1.20	1.32	1.62
Mass (mg)	1.62	1.93	3.64
Middle-ear Cavity			
Volume (mm ³)	358	455	678

B. FE modeling

A 3D FE mesh of the ear was generated in ANSYS. All components were meshed into eight-node hexahedral elements. Figure 1 shows the FE model of the pediatric ear including the external ear canal, middle ear and cochlea in

TABLE II. Dimensions of the pediatric ear model (cochlea) in comparison with published FE model of adult ear and the published data.

Structure	Pediatric model	Adult model (Zhang and Gan, 2013)	Published data
Cochlea			
Scala tympani volume (mm ³)	22.1	31.25	31.5 (Igarashi <i>et al.</i> , 1986)
Scala media volume (mm ³)	15.9	12.48	7.7 (Igarashi <i>et al.</i> , 1986)
Scala vestibuli volume (mm ³)	38.0	38.62	44.3 (Igarashi <i>et al.</i> , 1986); 29.2 (Thorne <i>et al.</i> , 1999)
BM			
Length (mm)	34.3	32.4	25.26–35.46 (Hardy, 1938)
RWM			
Area (mm ²)	1.34	2.34	2.29 (Okuno and Sando, 1988)

FIG. 1. (Color online) 3D finite element (FE) model of the pediatric ear from the anterior-medial view. The model consisted of the external ear canal, middle ear, and spiral cochlea. The middle ear included tympanic membrane (TM), three ossicles (malleus, incus, and stapes), two joints, suspensory ligaments and muscle tendons (C1, C2, C3, C4, C5, and C7), and stapedial annular ligament. The cochlea included the basilar membrane, Reissner's membrane, supporting structures, and three cochlear chambers (scala vestibule, scala tympani, and scala media). The middle-ear cavity and cochlear fluid are shown transparently. Explanations of C1, C2, C3, C4, C5, and C7 are given in Table III.

anterior-medial view. The middle-ear cavity and cochlear chambers are displayed with partial transparency. The external ear canal had a volume of 1037 mm³, and was meshed by 30 912 acoustic elements (Fluid 30 in ANSYS). The volume of the middle-ear cavity was measured as 358 mm³, and was divided into 28 920 eight-node hexahedral elements. The TM was meshed into two-layers of 2800 solid elements (Solid 185). The manubrium was represented by 216 solid elements. The malleus, incus, and stapes were meshed with 1820, 3030, and 120 solid elements (Solid 45), respectively.

For the cochlea, the BM and RM were meshed into 476 and 2023 solid elements, respectively. The volume of SV and ST was 38.0 and 22.1 mm³, respectively, and the volume of SM was 15.9 mm³. The perilymphatic fluid in SV and ST and the endolymphatic fluid in SM were modeled with eight-node hexahedral fluid elements (Fluid 80 in ANSYS). The SV and ST meshes contained 15 363 and 16 167 fluid elements, respectively, and the SM mesh contained 7378 elements.

To perform multiphysics analysis, two types of fluid-structure interactions were defined in acoustic elements of the ear canal and middle ear. Each surface next to a bony structure, such as the ear-canal wall and middle-ear cavity wall, was assumed to be fixed, and its acoustic element was represented by surface impedance. Each surface next to a movable structure, such as the TM and round window membrane (RWM), was defined in the model as a fluid-structural interface. The fluid motion within the cochlea was modeled based on the assumption that the fluid was incompressible. The velocity of each fluid node along any of the structural surfaces was equal to the corresponding velocity of the structure at that point.

To simulate MEE levels within the middle-ear cavity, the mesh of the entire middle-ear cavity was horizontally divided into several subdivisions. The fluid level filled in the cavity could be increased stepwise from the inferior bottom to fully fill the cavity. Corresponding to the fluid levels divided in the cavity, the TM was also meshed into different

sub-layers to match the corresponding fluid levels. The MEE in the cavity was assigned using eight-node hexahedral fluid elements (Fluid 80 in ANSYS). The bulk modulus of fluid was set as 2.2 GPa, the viscosity of fluid was set as 250 cp, which was in the range reported by [Carrie et al. \(1992\)](#). In this study, a fluid level at the location of the umbo was used for the model of the ear with MEE. The resulting volume of the MEE was about 160 mm³.

C. Material properties in the middle-ear model

Middle-ear soft tissues, which include the TM pars flaccida, TM pars tensa, incudomalleal joint, IS joint, stapedial annular ligament (SAL), and RWM, were modeled as linear viscoelastic materials. Other components of the middle ear and cochlea were modeled as elastic properties.

A linear viscoelastic model was used as the constitutive law for these six soft tissues, and the relaxation modulus of the tissue in time domain was expressed as ([Machiraju et al., 2006](#))

$$E(t) = E_0 + E_1 \exp\left(-\frac{t}{\tau_1}\right) \quad (1)$$

in which E_0 , E_1 , and τ_1 were viscoelastic parameters with constant values for each type of soft tissue, and t is the time. In the frequency domain, the linear viscoelastic behavior of material can be represented by the complex modulus,

$$E^*(f) = E'(f) + iE''(f), \quad (2)$$

in which $E'(f)$ was the storage modulus, $E''(f)$ the loss modulus, and f the frequency. For harmonic analysis, the relaxation modulus $E(t)$ in the time domain can be converted into a complex modulus $E^*(f)$ in the frequency domain by means of the Fourier transform. Based on the relaxation modulus in Eq. (1), the $E'(f)$, $E''(f)$, and loss factor $\eta(f)$ of the viscoelastic material were expressed as

$$E'(f) = E_0 + E_1 \tau_1^2 (2\pi f)^2 / (1 + \tau_1^2 (2\pi f)^2), \quad (3)$$

$$E''(f) = 2\pi E_1 \tau_1 f / (1 + \tau_1^2 (2\pi f)^2), \quad (4)$$

$$\eta(f) = \tan \delta = E''(f) / E'(f). \quad (5)$$

The storage modulus $E'(f)$ and loss factor $\eta(f)$ were specified for each type of soft tissue and were inputs to the FE model as the material properties.

Measured values of the mechanical properties of the soft tissues in pediatric ear are not currently available. Therefore, the viscoelastic parameters E_0 , E_1 , and τ_1 for each type of soft tissue were determined by a cross-calibration process using published measurements of EA in pediatric ears.

[Ruah et al. \(1991\)](#) examined the age-related morphologic changes of the human TM ranging from 2 days to 91 years using light and electron microscopy. They reported that the TM becomes less vascular, less cellular, more rigid, and less elastic with age. The mechanical properties of an adult human TM have been investigated in both experimental and modeling studies. The viscoelastic parameters E_0 , E_1 ,

and τ_1 of the TM used in an adult middle-ear FE model were 25 MPa, 70 MPa, and 25 μ s, respectively ([Zhang and Gan, 2013](#)). Viscoelastic parameters of other middle-ear soft tissues were reported in an adult-ear model ([Zhang and Gan, 2013](#)), which included the incudomalleal joint, IS joint, SAL, and RWM. These viscoelastic parameters of the adult tissues served as upper-limit values for the range of viscoelastic parameter values analyzed in calibrating the FE model of the pediatric ear.

The suspensory ligaments and ossicles were assumed to behave as isotropic elastic materials. The simulated responses to sound in the newborn ear ([Qi et al., 2008](#)) showed that varying the Young's modulus of each of the ossicles and middle-ear ligaments had little effect on TM volume displacement. Thus, the material properties of the suspensory ligaments and ossicles in the pediatric model were assigned the same values as those of the adult model reported by [Gan and Wang \(2007\)](#). Poisson's ratio was assumed to be 0.3 for the solid materials.

Except where noted below, the viscoelastic parameters of the middle-ear soft tissues in the normal ear were determined by the cross-calibration process ([Sun et al., 2002](#)) based on the EA measurements in young children with normal hearing ([Keefe et al., 2012](#); [Ellison et al., 2012](#)). These soft tissues are listed in the left column of Table III. The process target was to minimize the difference between the FE model and the measured EA data by selecting appropriate model parameters. One parameter was adjusted at a time for each tissue to analyze its effect on the FE model. Each model parameter was adjusted over a range with lower and upper limits equal to 0.2 and 1 times the value used in the adult model. The process was completed by minimizing the difference between the model results and the mean EA data reported by [Keefe et al.](#) and [Ellison et al.](#) The achieved mean percentage errors between the model results and measured data were less 12.5% over frequencies from 0.25 to 5 kHz.

After the viscoelastic parameters of the normal ear were determined, the behavior of the FE model in MEE ear was further validated against measurement data from two clinical otitis media groups reported by [Ellison et al. \(2012\)](#) and [Keefe et al. \(2012\)](#). The material properties of all middle-ear tissues and the cochlea in the MEE condition were the same as in normal ear model. The comparison between the model-derived EA in the MEE condition and the measured data from two clinical otitis media groups ([Ellison et al., 2012](#); [Keefe et al., 2012](#)) provided a validation of the model with different middle ear condition.

Maturational changes in ear-canal and middle-ear anatomy that influence the acoustical function of the ear were reviewed by [Abdala and Keefe \(2012\)](#). In newborn full-term infants up to age 2 months, the ear-canal wall moves in response to pressure changes in the ear canal ([Holte et al., 1991](#)). Evidence of functional immaturities involving ear-canal wall motion was inferred from acoustical ear-canal measurements of ER at ages 1 and 3 months using a model of ear-canal wall mobility ([Keefe et al., 1993](#)). These wall mobility effects become small by 3 months and the tympanic ring is fully ossified by 2 years ([Saunders et al., 1983](#)). While the ear-canal length and area continue to grow until

TABLE III. Mechanical properties of middle-ear soft tissues used for pediatric model in comparison with the published adult model.

Structure	Adult model	
	Pediatric model	(Zhang and Gan, 2011)
Tympanic membrane		
Pars tensa	E0 = 20 (MPa)	E0 = 25 (MPa)
Viscoelastic constant	E1 = 15 (MPa) $\tau_1 = 25$ (μ s)	E1 = 70 (MPa) $\tau_1 = 25$ (μ s)
Pars flaccida	E0 = 7 (MPa)	E0 = 7 (MPa)
Viscoelastic constant	E1 = 16 (MPa) $\tau_1 = 25$ (μ s)	E1 = 16 (MPa) $\tau_1 = 25$ (μ s)
Manubrium		
Elastic modulus	4700 (MPa)	4700 (MPa)
Incudomalleolar joint		
Viscoelastic constant	E0 = 60 (MPa) E1 = 180 (MPa) $\tau_1 = 20$ (μ s)	E0 = 60 (MPa) E1 = 180 (MPa) $\tau_1 = 20$ (μ s)
Incudostapedial (IS) joint		
Viscoelastic constant	E0 = 2.5 (MPa) E1 = 10 (MPa) $\tau_1 = 20$ (μ s)	E0 = 0.4 (MPa) E1 = 20 (MPa) $\tau_1 = 20$ (μ s)
Stapedial annular ligament (SAL)		
Viscoelastic constant	E0 = 1.4 (MPa) E1 = 5.0 (MPa) $\tau_1 = 20$ (μ s)	E0 = 2 (MPa) E1 = 10.8 (MPa) $\tau_1 = 20$ (μ s)
Superior malleolar ligament (C1)		
Elastic modulus (MPa)	4.9	4.9
Lateral malleolar ligament (C2)		
Elastic modulus (MPa)	6.7	6.7
Posterior incudal ligament (C3)		
Elastic modulus (MPa)	6.5	6.5
Anterior malleolar ligament (C4)		
Elastic modulus (MPa)	8.0	8.0
Posterior stapedial tendon (C5)		
Elastic modulus (MPa)	10	10
Tensor tympani tendon (C7)		
Elastic modulus (MPa)	8.0	8.0
Round window membrane		
Viscoelastic constant	E0 = 1 (MPa) E1 = 3 (MPa) $\tau_1 = 30$ (μ s)	E0 = 1 (MPa) E1 = 3 (MPa) $\tau_1 = 30$ (μ s)

adolescence, the ear-canal wall is substantially immobile at ages older than 2 years. Notwithstanding that fact, the effect of elasticity of the ear canal wall on EA was examined by varying the mechanical properties as described below.

To simulate the effects of a finite elasticity of the ear-canal wall on EA, a soft tissue layer with thickness of 1.2 mm was added to the ear canal of the model. The tissue properties were based on the measurements of Perry and Shelley (1955) over a range of tissue thicknesses from 1.0 to 1.5 mm. An acoustic-structural interface between the air and canal wall tissue was added to the FE model to couple the acoustic pressure and the structural displacement along the canal length. The canal wall tissue was meshed in three layers with a total of 8000 eight-node hexahedral solid elements. In this study, EA was predicted for values of the elastic modulus of the canal wall tissue ranging from 50 kPa to 2 MPa. The density of ear canal soft tissue was assumed to be 1000 kg/m³ between the density of water (1000 kg/m³) and that of undehydrated collagen (1200 kg/m³).

D. Model of passive cochlear mechanics

The cochlear model used in the present report was based on linear, passive cochlear mechanics. The model did not include the micromechanical structure of the organ of Corti, and active, nonlinear cochlear mechanisms including amplification were not considered. Nevertheless, a passive cochlear model described frequency-dependent interactions between the middle ear and cochlea through the oval and round windows. The linear, passive model described the longitudinal propagation of the traveling wave of BM deflection (Gan *et al.*, 2007). A middle-ear pressure gain was defined as the ratio of the pressure at the SV near the footplate to the pressure in the ear canal near the TM. This middle-ear pressure gain and related transfer functions between the ear canal and the stapes footplate motion were derived from the model.

The material properties of the cochlea components in the pediatric model were assigned the same values as in an adult FE model with a spiral cochlea (Zhang and Gan, 2011). The elastic modulus of the BM was varied from 50 MPa at the base to 3 MPa at the apex. A material-dependent Rayleigh-type damping in ANSYS was specified for the BM. The material-dependent Rayleigh damping expressed the element damping matrix (C) as $C = \beta K$, and K is element stiffness. In the present study, the damping coefficient β (or stiffness damping) was assumed to increase linearly from 0.2×10^{-3} s at the base of the BM to 0.1×10^{-2} s at the apex. The bulk modulus of the cochlear fluid was set as 2.2 GPa as in a previous adult study (Zhang and Gan, 2013).

E. Calculations of EA, middle-ear and cochlear transfer functions

The EA of the FE model was calculated from the acoustic impedance in the ear canal (Stinson *et al.*, 1982; Keefe and Feeney, 2009). Briefly, a uniform normal velocity was applied at a reference location in the ear canal 20 mm from the TM. The sound pressure in the ear canal was derived based on analysis of fluid–structure coupled interaction, and the input impedance Z_{EC} at this reference location was defined as the ratio of the mean sound pressure (P_c) to the volume velocity at this location, i.e.,

$$Z_{EC} = \frac{P_c}{U_0 \cdot A_0}, \quad (6)$$

in which U_0 was the normal particle velocity and A_0 the cross-sectional area of the ear canal, with both variables defined at the reference location. The acoustic pressure reflectance $R_p(f)$ at this reference location was computed from the normalized ear canal impedance as

$$R_p(f) = \frac{\frac{Z_{EC}}{Z_c} - 1}{\frac{Z_{EC}}{Z_c} + 1}, \quad (7)$$

in which $Z_c = \rho c / A_0$ was the characteristic impedance of the ear canal at this location, ρ the density of air, and c the phase velocity of sound in air. In this paper, the normal velocity

excitation (U_0) was applied at the reference location, and sound pressure (P_c) was extracted in the same location. During the reflectance calculation of this study, A_0 was eliminated from Eq. (7) because both Z_{EC} and Z_c include A_0 .

The energy reflectance, ER , was calculated from $R_p(f)$ by

$$ER(f) = |R_p(f)|^2. \quad (8)$$

The energy absorbance, EA , as a function of frequency was

$$EA(f) = 1 - ER(f). \quad (9)$$

Middle-ear and cochlear transfer functions further described below were calculated from the model with normal-ear and MEE conditions across the frequency range of 0.2 to 8 kHz.

To calculate the displacement of TM, stapes footplate, and BM, the sound pressure was applied at 2 mm away from the TM in the canal. This is a similar loading location to that used in human temporal-bone measurements of the middle-ear transfer function. The vibrations of the TM, stapes footplate, and BM were calculated in response to this sound pressure.

The relative level changes in TM, BM displacement, and EA in a pediatric ear with MEE were calculated relative to the corresponding responses in a normal ear. The relative level changes ΔD in TM and BM displacement were transformed to a logarithmic scale in decibels and calculated as

$$\Delta D = 20 \log_{10} \left(\frac{D_{MEE}}{D_N} \right). \quad (10)$$

The EA level change ΔEA due to MEE was represent as a logarithmic scale in decibels and calculated as

$$\Delta EA = 10 \log_{10} \left(\frac{EA_{MEE}}{EA_N} \right), \quad (11)$$

in which the subscripts N and MEE represent the corresponding response in the pediatric ear with normal hearing and MEE, respectively. The EA level representation in Eq. (11) allows for useful comparisons to other logarithmic measures of hearing function (Allen *et al.*, 2005).

The FE analysis ran on a personal computer (Dell Precision T7910, 32 GB memory) using ANSYS 15.0, and the computation time was 18 min, 20 s for all responses at 20 frequencies.

IV. RESULTS

A. FE model-derived EA curves of normal ear

The model was first compared with the published clinical data measured in normal ears. The viscoelastic material properties that were determined for the pediatric FE model through the cross-calibration process are listed in Table III with material property values of an adult FE model (Zhang and Gan, 2013) for comparison.

Figure 2 shows the predicted EA in comparison with the clinical measurements in young children (Keefe *et al.*, 2012; Ellison *et al.*, 2012) across the frequency range of 0.2 to 8.0 kHz. The predicted EA was 0.158 at 0.4 kHz, increased to 0.413 at 1.2 kHz, and reached a maximum of 0.90 at 4.0 kHz. The EA decreased as frequency further increased and reached 0.214 at 8.0 kHz. There were two peaks in the predicted EA: a small peak around 1.2 kHz and larger peak around 4.0 kHz. The predicted EA was generally consistent with the mean EA in Keefe *et al.* Compared with the range of EA responses measured by Ellison *et al.*, the EA in the model was within the inter-quartile range (IQR) of measurements below 4 kHz and slightly larger than the 75% of measurements above 4 kHz.

B. Evaluation of pediatric model for otitis media

The pediatric FE model was validated by simulating the MEE ear (of age 4 years) against measurement data of two clinical otitis media groups reported by Ellison *et al.* (2012) and Keefe *et al.* (2012). Figure 3 shows the predicted EA in comparison with the clinical measurements in the two otitis media groups by Keefe *et al.* (2012) and Ellison *et al.* (2012) across the frequency range of 0.2 to 8.0 kHz. The predicted EA generally increased with increasing frequency in the range below 4.4 kHz: EA was 0.155 at 0.25 kHz, 0.231 at 0.8 kHz, and a maximum of 0.610 at 4.4 kHz. EA decreased at higher frequencies to 0.117 at 8 kHz. The predicted EA had one peak near 0.4 kHz and another near 4 kHz. The predicted EA of the ear with MEE was within the 25% to 75% range of measurements of Ellison *et al.* except for frequencies below 1.2 kHz. The predicted and mean EA from Keefe *et al.* (2012) were generally similar across frequency, although the frequency of the maximum EA was higher in the model (4 kHz) than the measured data (2.8 kHz).

To compare the model for normal hearing and MEE, Fig. 4 displays EA predicted by the pediatric model under both normal and MEE conditions. In agreement with measurements, EA was significantly reduced in the MEE ear. The

FIG. 2. (Color online) FE model-derived energy absorbance (EA) of pediatric ear in normal condition compared with the published measurement data in normal ears. The thin dotted lines represent the inter-quartile range (IQR) of measurements by Ellison *et al.* (2012) from 59 ears. The thick line with square symbols represents the mean EA measurement in 43 ears by Keefe *et al.* (2012) for the normal group. Each error bar denotes ± 1 standard error (SE) of the mean EA.

FIG. 3. (Color online) FE model-derived EA curve for the ear with middle-ear effusion (MEE) compared with published EA data measured in pediatric patients with otitis media. The thick dashed line with triangle symbols represents the median EA measurements by [Ellison et al. \(2012\)](#) with the 25% to 75% range plotted in dotted lines. The thick line with circle symbols represents the mean EA reported by [Keefe et al. \(2012\)](#) for children with otitis media who have a conductive hearing loss (based on an air-bone gap of 20 dB or more at 0.5 kHz). The vertical error bars denote ± 1 SE from the mean EA.

predicted EA at a small peak near 1.2 kHz in the normal ear was reduced in amplitude from 0.413 in the normal ear to 0.110 in the MEE ear (with no peak in that frequency region). MEE caused a significant reduction of EA at all frequencies above 0.6 kHz. The difference of EA between the normal and MEE ears was 0.193, 0.356, and 0.370 at a frequency of 1, 3, and 5 kHz, respectively.

C. Middle-ear transfer function derived from the model

Figure 5 shows the predicted pediatric transfer functions of the umbo and stapes footplate (FP) displacements relative to ear-canal pressure just in front of the TM. As a comparison of modeling results between adult and pediatric FE models, the published adult transfer functions from [Zhang and Gan \(2011\)](#) are also included.

For the pediatric model, the umbo displacement transfer function magnitude was relatively constant near $0.11 \mu\text{m}/\text{Pa}$ between 0.2 and 1.2 kHz and decreased at frequencies above 3.2 kHz. At about 1.2 kHz the umbo displacement of the pediatric model showed a resonance peak. The stapes FP

FIG. 4. (Color online) Comparison between the FE model-derived EA curves for the normal ear and the ear with MEE.

FIG. 5. (Color online) The transfer functions of umbo displacement relative to ear-canal pressure just in front of the TM, and of stapes FP displacement relative to the same ear-canal pressure, are plotted for the pediatric FE model and an adult FE model ([Zhang and Gan, 2011](#)). Also plotted is the transfer function of the TM displacement relative to the ear-canal pressure just in front of the TM (solid line with filled squares) from the pediatric model. (a) Magnitude in $\mu\text{m}/\text{Pa}$, (b) phase in degrees.

displacement transfer function magnitude was relatively constant near $0.022 \mu\text{m}/\text{Pa}$ at frequencies between 0.2 and 1.2 kHz. These magnitudes in the pediatric model were similar to the corresponding magnitudes in the adult model up to above 1.2 kHz, while the magnitudes at higher frequencies over about 1.5 kHz rolled off more steeply for the pediatric than the adult model.

A mean TM displacement was defined in the pediatric FE model as the ratio of the total volume displacement of the TM to the surface area of the TM. The magnitude of the umbo displacement transfer function was slightly larger than the magnitude of the mean TM displacement transfer function at frequencies below 1.5 kHz, and smaller at frequencies above 3.0 kHz. The largest differences between the magnitudes of the umbo and mean TM displacement transfer functions occurred at frequencies above 4.0 kHz. The phase of the mean TM displacement transfer function lagged less than the phase of the pediatric umbo displacement transfer function over frequencies from 0.2 to 8.0 kHz.

The middle-ear pressure gain was defined as the level of the magnitude of the ratio of the cochlear pressure in the SV near the stapes footplate to the ear-canal pressure at the surface of the TM. This transfer function characterized the transformation of sound from the ear canal to the cochlea through the middle ear. Figure 6 displays the predicted middle-ear pressure gain in comparison with the published

FIG. 6. (Color online) Middle-ear pressure gain derived from the pediatric model (solid line with open circles) and compared with the mean experimental results from the adult temporal bones by Aibara *et al.* (2001) (dashed line) and Nakajima *et al.* (2008) (solid line).

measurements by Aibara *et al.* (2001) and Nakajima *et al.* (2008). The mean Aibara *et al.* data were measured from 11 adult temporal bones in which the cochlear pressure was detected using a hydrophone. The mean Nakajima *et al.* data were measured from 6 temporal bones, in which the cochlear pressure in SV at the base of the cochlea was measured at an approximate distance of 200 μm from the surface of the cochlea using a fiber-optic pressure sensor. Over most frequencies, the results from the pediatric model were similar to those experimental data from the temporal bones. The model-predicted maximum middle-ear pressure gain was approximately 23 dB near 1.2 kHz.

D. BM vibration derived from the model

Given the assumption that the cochlear mechanics were linear and passive, the pediatric FE model predicted BM responses as a function of location along the BM from its base to its apex. Figure 7 shows the FE model-derived best-frequency cochlear map along with the cochlear frequency-position function developed empirically by Greenwood (1990) for an average human adult ear. This map indicates the frequency corresponding to the peak BM vibration as a function of location along the BM. As shown in Fig. 7, the slope of the modeling curve was larger than Greenwood's

FIG. 7. (Color online) Simulated best-frequency map and compare with experimental data by Greenwood (1990).

results. At the location of 16 mm from the base, the best frequency shifted from the experimental value of 1.85 kHz to the modeling value of 1.6 kHz. This shift is more likely due to individual variability and/or modeling uncertainties rather than any systematic difference between the cochlear frequency-position maps of an adult ear compared to the ear of a 4-year-old.

Figure 8 displays the simulated transfer function level of BM displacement (d_{BM}) as a function of location along the BM relative to the displacement of the stapes FP (d_{fp}) for sinusoidal stimulus frequencies between 0.4 and 8.0 kHz. These simulated responses were generated by the pediatric FE model with linear, passive cochlear mechanics. Each such normalized, single-frequency BM displacement transfer function illustrates the mechanical tuning along the BM.

The cochlear parameters in the model were adjusted such that the predicted BM displacement transfer function was similar in level to mean measured data in human temporal bones at basal locations near tonotopic frequencies of 2.0 kHz (Stenfelt *et al.*, 2003). Stenfelt *et al.* measured BM responses in the basal turn of 12 mm from the RM, and the BM vibration normalized with the velocity of the stapes footplate was about 25.5 dB (from Fig. 5 in Stenfelt *et al.*, 2003). The BM vibration predicted at 2.0 kHz was 26.5 dB at a tonotopic peak location 13 mm from the basal turn.

The predicted BM response curves tended to become sharper and narrower with increasing frequency (Fig. 8). The peak amplitude ratio was around 23 dB at frequencies above 0.8 kHz. The peak values were also similar to the other numerical results (Elliott *et al.*, 2013).

These results based on the assumed maturity of cochlear mechanics by age 4 years are supported by the experimental evidence of cochlear maturation reviewed in Abdala and Keefe (2012), and of the maturation by 2 years of age of such electrophysiology measures of auditory function as the auditory brainstem response, middle latency response, the late cortical P2 component of the obligatory cortical evoke potential, and the mismatch negativity (Eggermont and Moore, 2012). Such a maturity of peripheral electrophysiological responses in the auditory system is substantial evidence that the cochlear mechanics are mature by age 2 years.

FIG. 8. (Color online) Predicted transfer function of BM displacement to stapes FP displacement by the pediatric FE model at each BM location for a sinusoidal sound stimulus.

E. Predicted effects of MEE on ear-canal, middle ear and cochlear function

The pediatric FE model was used to compare the effects of MEE relative to normal function on responses at locations in the ear canal, middle ear and cochlea. Figure 9 shows the relative level changes caused by MEE in EA, umbo displacement, and BM displacement relative to the values predicted in a normal ear. The relative level change of displacements and EA were calculated using Eqs. (10) and (11), respectively. The negative value indicates that the MEE resulted in a reduction of the response. The relative level difference of EA across frequency was larger on average than that of the umbo and the BM displacement curves, and had a smaller variation over frequency. The EA level showed a slight increase at low frequencies and a reduction above 0.4 kHz. In contrast, umbo and BM displacement levels showed larger reductions over a wider frequency range. The relative level differences in EA, umbo displacement, and BM displacement decreased with increasing frequency up to a minimum level at 1.2 kHz, with maximum reductions of 6.1, 11.5, and 14.5 dB, respectively. The MEE reductions of EA and umbo displacement were within around 5 dB at all frequencies below 6 kHz, except near 1.2 kHz. Above 6 kHz, the relative level change in EA was within a couple dB, whereas the relative level change in umbo displacement decreased more than -20 dB at 8 kHz. The overall level reduction due to MEE for BM displacement level was larger from about 0.6 to 3 kHz compared to umbo displacement level and EA.

F. Effect of ear canal elasticity on EA

As described in Sec. III of “Finite Element Model,” the main pediatric FE model was constructed under the assumption that the ear-canal wall was rigid by age 4 years. Notwithstanding that fact, a further model analysis was performed to examine the effect on EA of varying the ear-canal elasticity. The results in Fig. 10 show six model-derived EA curves with different values of the elastic modulus of the canal wall tissue in the range from 50 kPa up to a rigid-wall condition. Results were calculated in the normal-ear pediatric model for two material damping (β) values. At frequencies below 4 kHz, a lower elastic modulus generally resulted

in a larger EA for both damping values, except that EA for the softest tissue (50 kPa) was smaller than EA for the tissue with an elastic modulus of 100 kPa at frequencies in the range from 2 to 3.8 kHz. Above 4 kHz, there was a steeper roll-off in EA for larger values of the elastic modulus. Comparing the curves in Figs. 9(a) and 9(b), the larger value of the canal-wall damping coefficient β produced larger changes in the predicted EA across the range of values of the canal-wall elastic modulus, particularly at frequencies below 4 kHz.

The predicted EA curves converged to the EA for the rigid canal wall with increasing values of the elastic modulus of the canal wall. When the elastic modulus was 2 MPa, the canal wall elasticity had only a negligible effect on EA compared to the rigid-wall condition. Overall, the lower elastic modulus and higher damping of the ear canal wall increased the predicted EA. This increase was a result of additional energy loss in the soft ear canal. If an assumed tissue thickness of canal wall exceeded 1.2 mm, it would produce even larger effects on EA at low frequencies (these results are not shown). The ear canal of the infant at age less than one year has a much softer canal wall than the ear of an adult or 4-year-old. A new model would be more appropriate to study the effect of ear-canal wall elasticity properties on sound transmission through the middle ear of a young infant.

FIG. 9. (Color online) Comparison of the predicted relative level changes of EA, umbo displacement level, and BM displacement level in the MEE condition relative to the normal-ear condition.

FIG. 10. (Color online) Effect of varying the material properties of ear-canal soft tissue on EA. (a) EA predictions with varying elastic moduli for damping parameter $\beta = 0.1 \times 10^{-4}$ s. (b) EA predictions with varying elastic modulus for damping parameter $\beta = 0.2 \times 10^{-4}$ s. The model curve in the legend corresponds to the limit of rigid ear-canal walls.

V. DISCUSSION

A. Effects of MEE on EA, middle-ear and cochlear transfer functions

The presence of fluid within the middle-ear cavity increased the mass and stiffness of the middle-ear system, and affected sound transmission through the ear in terms of the relative level changes in EA, and in the middle-ear and cochlear transfer functions. The level changes in Fig. 9 due to the presence of MEE were greater for all measures between 1 and 2 kHz than at other frequencies. This indicates the importance of measuring EA and umbo displacement at these frequencies to screen for MEE in young children.

In Fig. 9, the reduction level in EA due to MEE was less than 6 dB, and the largest reduction of BM displacement was about 14.5 dB. The EA, umbo and BM displacement level changes provide different views of how the MEE impedes the sound transmission from the ear canal into cochlea. EA is one of power-flow variables that can provide useful information on the status of the ear canal and middle ear (Allen *et al.*, 2005). Unlike EA, which measures the response of the whole TM, umbo displacement is a more direct measure of the dynamic state of the middle ear ossicles and is less influenced by the compliance of the TM (Rosowski *et al.*, 2012), which is a possible explanation of the more correlation of the umbo displacement level loss with the BM loss.

A larger MEE-induced reduction in BM displacement than in TM displacement or EA may have resulted from the following possible factors: (1) MEE increased the middle-ear mass, and thereby the resonance frequency of middle ear was shifted towards lower frequencies, resulting in the change of umbo displacement peak. In the meantime, the increase of middle ear stiffness due to MEE also reduced the TM movement in low frequencies (Ravicz *et al.*, 2004). The reduction of TM displacement may result in a decrease of BM displacement level via reduced energy transfer through the ossicular chain. (2) The presence of MEE changed the pattern of TM motion (Zhang and Gan, 2013), and such a change might influence the vibration transmission through the ossicular chain. This in turn would alter the energy transmitted to the cochlea. (3) The presence of fluid within the middle-ear cavity in the MEE condition might change the interaction at the round-window membrane relative to the normal condition of air within the middle-ear cavity. The presence of fluid restricted the motion of the round-window membrane and reduced the BM displacement level.

Group studies in children show that EA is reduced across a wide frequency range in ears with conductive hearing loss (CHL) compared to normal-hearing ears, but that the reduction in EA or EA level is not proportional to the amount of CHL (Keefe *et al.*, 2012). For example, Keefe *et al.* concluded that EA accurately classified ears as having either normal hearing or a CHL, but EA did not predict the frequency of the CHL, when a CHL was present. Moreover, they reported only slight differences in EA in impaired ears with varying criterion values of CHL (e.g., exceeding 20, 25, or 30 dB air-bone gap at any octave frequency between 0.25 and 4 kHz, or any of the five octave frequencies), whereas there were substantial differences in any of these groups of

impaired ears compared to the normal group. Thus, differences in EA levels across frequency do not predict the degree of CHL, but only whether a CHL is present. This explains why clinical decision theory assessing the ability of ER or EA to classify ears as normal or impaired has often been used, rather than analyses to regress the amount of a CHL based on EA or some related acoustic transfer function in the ear canal. This is the basis for the recommendation to refer a child for additional audiological and medical evaluations on the basis of evidence from an EA test of increased risk for having CHL (Keefe *et al.*, 2012), and similarly for increased risk of MEE (Ellison *et al.*, 2012).

In addition, the relative magnitude of the BM shifts were only the order of 15 dB or less. This contrasts with the pediatric air-bone gap data in Keefe *et al.* (2012), which defined groups of CHL ears based on losses larger than 20 or 30 dB. Most of this CHL arose as an elevation in the air-conduction audiogram, although Keefe *et al.* did not require CHL ears to have bone conduction audiograms within normal limits. One reason for the smaller reduction in BM displacement level compared to the air-bone gap data was the likely smaller volume of MEE in the model. Rovers *et al.* (2004) reported that otitis media with effusion causes change of hearing levels from 10 to 40 dB based on the volume of MEE. In the present model, the MEE level were assumed to be at the location of the umbo. A further increase of the MEE level would result in a larger reduction in the predicted displacement levels of the umbo and BM. Another reason was that the model only included the effect of the MEE, the potential change of middle ear pressure due to otitis media with effusion were not reflected in the model. Therefore, the predicted FE-model change does not appear as large as the conductive attenuation in the ears of children with OME. This remains a topic for future research.

B. Effect of geometry and mechanical properties on EA

Developmental variations in the geometry and mechanical properties of the middle ear affected the EA. To separately identify the effects of the mechanical properties and geometry on EA, the EA was predicted for three cases: a normal pediatric model, an adult model, and a “stiffened pediatric model.” The stiffened pediatric model used the same mechanical properties of the ear tissues as for the adult model (as listed in Table III for adult), but used geometrical parameters corresponding to those of the normal pediatric ear model.

Thus, the differences between the stiffened pediatric model and the adult model identified effects of the physical size or geometry of the ear on the predicted EA. The differences between the stiffened pediatric model and the normal pediatric model identified effects of maturational differences in mechanical parameters on EA.

Figure 11 displays the EA curves obtained from these three cases, in which the EA curve for the normal pediatric model was the same as in Fig. 3. The adult EA was higher than the normal pediatric EA at 0.6 to 2 kHz but lower at frequencies over 2 kHz. Particularly, at frequencies below

FIG. 11. (Color online) The predicted EA for the normal pediatric ear model, adult ear model, and stiffened pediatric ear model.

0.6 kHz, the EA was almost the same for pediatric and adult ears.

Below 0.6 kHz, EA was slightly larger for the stiffened pediatric model than for the other two models. This suggests that the maturational effects of geometrical and mechanical parameters offset one another. At frequencies from 0.6 to 3.5 kHz, EA for the stiffened pediatric model was intermediate in value between EA for the normal pediatric and adult models. The EA curve from the stiffened pediatric model was the same as from the adult model above 5 kHz. This suggests that there were no maturational differences in geometry affecting EA above 5 kHz.

The comparison across frequency between EA for the normal pediatric and stiffened pediatric models suggests that mechanical properties may have affected EA at frequencies above 0.6 kHz and particularly at frequencies from 0.8 to 4 kHz. An increase of tissue stiffness resulted in a flattened EA curve.

Comparing the EA in the normal pediatric and stiffened pediatric models, EA was larger in the normal pediatric model above 5 kHz. This suggests that maturational differences in mechanical parameters were responsible for this difference at high frequencies. These differences would affect the stiffness matrix in the FE model.

Beers *et al.* (2010) compared normal pediatric ER data (average age of 6.15 years) with normal adult ER data obtained by Shahnaz and Bork (2006). The results of Beers *et al.* showed that ER was larger (corresponding to a smaller EA) in ears of Caucasian children than in adults at frequencies below 2 kHz. The present model results for EA showed general consistency with the measured results except below 0.4 kHz.

Apart from the effect of geometry and mechanical properties of the middle ear on EA, the sensitivity analysis in Fig. 10 also shows EA was sensitive to the mechanical properties of the ear canal. As a general summary for the results in Figs. 10 and 11, the physical size or geometry of the middle ear influenced EA in the low- and mid-frequency range, and the mechanical properties of outer and middle ear soft tissues had an effect on EA at most frequencies.

From both research and clinical perspectives, it is important to distinguish the variations in EA measurements and determine factors caused by developmental process or

pathology of the ear. Otitis media is the most commonly diagnosed disease in young children. The pediatric model reported in this study provided a more realistic FE model to study sound transmission differences between ears with normal function and ears with otitis media, and to study the mechanisms that lead to hearing loss in many ears with otitis media. More research is needed to construct FE models to predict EA in children younger than 4 years old.

VI. CONCLUSION

In this study, a pediatric FE model was developed to predict middle-ear and cochlear functions in normal and otitis media with effusion ears, and compared to predictions in normal ears using adult FE models. This pediatric model consisted of an ear canal, middle ear, and spiral cochlea based on histological section images of a temporal bone of a 4-year-old child. FE model represented the middle-ear and passive cochlear mechanics of the pediatric ear in terms of a specification of the anatomical structure of the ear and the mechanical properties of ear tissues. The mechanical property parameters of the middle-ear tissues were determined by a cross-calibration process based on published EA measurements in young children with normal middle-ear function. The model was used to simulate the effects of MEE on EA; these predictions were generally similar to clinical EA measurements in ears with MEE. The relative level changes in umbo motion and BM displacement of a pediatric ear with MEE were also predicted using the model. The results showed the relative level change of EA had less sensitivity to the MEE than those of umbo and BM motions. This study provides the first 3D FE model of a pediatric ear that was validated using clinically measured EA data. The model has potential applications for investigating the mechanisms of otitis media, which is the most common middle-ear disease in young children.

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