Hearing Research 349 (2017) 111-128

Contents lists available at ScienceDirect

Hearing Research





**Review Article** 

## Analytical and numerical modeling of the hearing system: Advances towards the assessment of hearing damage



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#### ARTICLE INFO

Article history: Received 29 June 2016 Received in revised form 20 January 2017 Accepted 25 January 2017 Available online 2 February 2017

Keywords: Auditory system modeling AHAAH model Finite element model Hearing damage modeling Cochlea micromechanics modeling

## ABSTRACT

Hearing is an extremely complex phenomenon, involving a large number of interrelated variables that are difficult to measure in vivo. In order to investigate such process under simplified and well-controlled conditions, models of sound transmission have been developed through many decades of research. The value of modeling the hearing system is not only to explain the normal function of the hearing system and account for experimental and clinical observations, but to simulate a variety of pathological conditions that lead to hearing damage and hearing loss, as well as for development of auditory implants. effective ear protections and auditory hazard countermeasures. In this paper, we provide a review of the strategies used to model the auditory function of the external, middle, inner ear, and the micromechanics of the organ of Corti, along with some of the key results obtained from such modeling efforts. Recent analytical and numerical approaches have incorporated the nonlinear behavior of some parameters and structures into their models. Few models of the integrated hearing system exist; in particular, we describe the evolution of the Auditory Hazard Assessment Algorithm for Human (AHAAH) model, used for prediction of hearing damage due to high intensity sound pressure. Unlike the AHAAH model, 3D finite element models of the entire hearing system are not able yet to predict auditory risk and threshold shifts. It is expected that both AHAAH and FE models will evolve towards a more accurate assessment of threshold shifts and hearing loss under a variety of stimuli conditions and pathologies.

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http://dx.doi.org/10.1016/j.heares.2017.01.015

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ABM	Artificial Basilar Membrane	N
AC	Air Conduction	0
AHAAH	Auditory Hazard Assessment Algorithm for Humans	0
AR	Acoustic Reflex	0
ARU	Auditory Risk Unit	0
BC	Bone Conduction	P
BM	Basilar Membrane	P
CFD	Computational Fluid Dynamics	R
СР	Cochlear Partition	R
CTS	Compound Threshold Shift	R
DRC	Damage Risk Correlate	S
EABR	Evoked Auditory Brain Responses	S
FE	Finite Element	S
HC	hair cells	S
ICE	Integrated Cochlear Energy	Te
IHC	Inner Hair Cells	T
LIG	Spiral Ligament	Т
		W

MEM-FSS Micro-Electro-Mechanically based Frequency			
Signature Sensor			
Micro-Electro-Mechanical Systems			
Organ of Corti			
Outer Hair Cells			
Osseous Spiral Lamina			
Oval Window			
Phalangeal Supporting Cells			
Permanent Threshold Shift			
Reticular Lamina			
Reissner's Membrane			
Round Window			
Scala Media			
Sound Pressure Level			
Scala Tympani			
Scala Vestibuli			
Tectorial Membrane			
Tympanic Membrane			
Travelling Wave			
Wentzel-Kramers-Brillouin-Jeffreys			

## 1. Introduction

Hearing is an extremely complex phenomenon, involving a large number of interrelated variables that are difficult to measure in vivo. Much of our understanding of the hearing process has emanated from experimental and clinical research studies. However, an important complement to clinical observations and animal model research is the development of analytical and numerical models of the auditory system. Modeling is essential for the study of those aspects of hearing that are difficult to unveil from experiments. A model may be used to determine the basic mechanisms behind the hearing process and to investigate a variety of pathological conditions that lead to hearing loss. In order to study such process under simplified and well-controlled conditions, models of sound transmission have been developed through many decades of research.

High intensity noise (e.g. blast exposure in the military) over 170 dB SPL can result in otologic injury and reduced hearing. In addition to studies of the normal hearing process, research studies on hearing damage are essential to better understand, prevent and counteract the effects of the many factors contributing to hearing loss. However, conducting quantitative assessment studies of hearing damage is challenging for many reasons. For instance, inducing hearing damage in vivo, either in animal models or humans, raises ethical concerns and it is limited by practical considerations. Modeling, again, is an important approach that can help to better understand the risk of loss of hearing function following high intensity noise exposure, and for the development of hearing personal protective equipment and other auditory hazard countermeasures.

In this review, we intent providing the general reader, and the modeler in particular, with a broad view of the strategies used to model the hearing system in its physiological and pathological conditions. While space limitations will not allow us to discuss all the manuscripts addressing modeling in the auditory field, this review summarizes some of the seminal works on analytical, physical and finite element (FE) modeling, from the first half of the 20th century until the most recent and significant developments at the present time. In section 2, we describe lumped parameter models and finite element models of the outer ear function. In

section 3, we focused on the key modeling studies leading to our current understanding on the behavior of the tympanic membrane (TM) and the middle ear, comprising analytical, numerical and lumped parameter models. The inner ear has been the object of most modeling studies, since it is the location where the mechanotransduction process occurs, transforming the sound stimulus into electrical impulses. Modeling of the inner ear is covered in section 4, and it includes physical models, analytical models of cochlea hydrodynamics and basilar membrane (BM) vibration and finite element models of the cochlea. More refined and specific micromechanical and mechanoelectrical models of the organ of Corti (OC) are summarized in section 5. The following section 6, is dedicated to the few lumped parameter models and finite element models that have integrated the behavior of the whole hearing system. Of particular significance to this review is the summary of key aspects involved in hearing damage presented in section 7. While these models are still scarce, the numerical and analytical models of the outer and middle ear presented in this section have recently extended our knowledge of hearing damage due to high intensity sound on this ear compartments. Also included here are the nonlinear modeling efforts of the cochlea and the outer hair cells (OHC) in the OC. We close section 7 with a description of the evolution of the Auditory Hazard Assessment Algorithm for Human (AHAAH) model currently used in the military for prediction of hearing damage due to high intensity sound pressure. A summary of the progress on modeling of the hearing system and the status of modeling of hearing damage is presented in the last section of this review.

## 2. Outer ear models

## 2.1. Outer ear lumped parameter models

The ear canal has been modeled using lumped parameter approaches to investigate the overall effect of this anatomical structure on sound conduction to the tympanus. At low frequencies, the ear canal can be considered as a simple rigid-walled cavity characterized just by its volume. In a circuit model, this can be corrected for in the experimental data (Zwislocki, 1957) or it can be explicitly represented by just a single capacitance. At higher frequencies the wavelength starts to become comparable to the canal length and standing-wave patterns start to form along the canal. As a first approximation this can easily be modeled as a single mass-spring combination to produce the first natural frequency (Onchi, 1949, 1961) or the canal can be modeled analytically as a uniform transmission line (Wiener and Ross, 1946). A one-dimensional modified horn equation can be used to model the effect of the non-uniformity of the transverse canal dimensions (Khanna and Stinson, 1985) as well as the effect of the distributed acoustical impedance of the tympanic membrane (Stinson and Khanna, 1989). Many experimental studies have been conducted to determine the ear canal impedance and reflection coefficient (Keefe et al., 1993; Voss and Allen, 1994), the sound power per unit cross-sectional area (Farmer-Fedor and Rabbitt, 2002) and the pressure level generated at the entrance to the ear canal at six azimuthal angles of the incident sound (Shaw, 1966). Stinson (1985) solved the modified horn equation and predicted the pressure distribution along the ear canal curvature. Kahana and Nelson (2006, 2007) generated an accurate mesh from the head, the auricle, and the external ear geometry using a 3D laser scanner technique. The frequency response of each model was obtained by solving the Helmholtz equation using the boundary element method. Stinson and Daigle (2005) calculated the sound field arising along the central axis and the cross sections in a detailed model of the human ear canal using two approaches: an analytic horn equation and a boundary element method for prediction of the sound pressure magnitude along the curvature of the ear canal at different frequencies.

### 2.2. Outer ear numerical models

The effect of non-uniform transverse canal dimensions and distributed acoustical impedance of the tympanic membrane have also been investigated using coupled mechanical and acoustical finite-element models. For example, Koike et al. (2002) compared effects of a more or less realistic canal shape (curved, but with the eardrum not tilted in a realistic way) with those of a simple cylindrical canal. At 7 kHz, the variation of pressure across the eardrum was less than 2 dB. At even higher frequencies, the wavelength becomes comparable to the transverse canal dimensions and the pressure starts to be non-uniform across the canal (Stinson and Daigle, 2005). At very high frequencies, the precise orientation and shape of the canal termination at the tympanic membrane also become important. Rabbitt and Holmes (1988) modeled this effect analytically using asymptotic approximations. Tuck-Lee et al. (2008) used a special adaptive finiteelement approach to facilitate the calculations at high frequencies. In a finite element model of the human ear canal, Hudde and Schmidt (2009) found acoustical modes that raise interesting questions about the notion of a midline axis that is often assumed in canal modeling. Although Tuck-Lee et al. (2008) did allow for energy absorption in the walls of the canal, most models have assumed that the canal walls are perfectly rigid. However, for the newborn human canal, in which the bony canal wall has not vet formed. Oi et al. (2006) found that the response of a hyperelastic finite-element model to large static pressures (as used in tympanometry) and to auditory frequencies (Gariepy, 2011) is strongly affected by the compliance of the canal wall. In most ear-canal modeling, the sound source is characterized by sound-pressure measurements with a probe microphone inside the canal, which can have significant effects in some circumstances (Zebian et al., 2012). In modeling the response of the ear to free-field sound with the microphone outside the canal, it is necessary to take into account the shape of the pinna (Hudde and Schmidt, 2009) and perhaps even the shape and dimensions of the whole head.

#### 3. Tympanic membrane and middle ear models

#### 3.1. Analytical and lumped parameter models

The first quantitative model of tympanic membrane function was the "curved-membrane" hypothesis of Helmholtz (1868). This was a distributed-parameter analytical model that depended critically on the curvature of the tympanic membrane. It also made use of anisotropy, assuming differences between the radial and circumferential directions. The model was further developed by Esser (1947) and by Guelke and Keen (1949), and an error in Helmholtz' calculations was pointed out by Hartman (1971). To address spatial patterns of the tympanic membrane, analytical models were proposed by Frank (1923), Gran (1968), and Wada and Kobayashi (1990), but all were forced to make many oversimplifications, including the critical one of taking the tympanic membrane as flat rather than conical. Asymptotic analytical models have been more successful (Rabbitt and Holmes, 1986; Fay, 2001). Goll and Dalhoff (2011) recently presented a 1-D string model of the tympanic membrane that can be viewed as a distributed variant of the lumped delay-line models. In the circuit model of Onchi (1949, 1961), the human tympanic membrane was represented by a single mass attached by springs to the tympanic annulus and to the manubrium. Zwislocki (1957) and Møller (1961) represented the human tympanic membrane with two parts, one branch in parallel with the ossicular chain, corresponding to the compliance and the resistance of the eardrum when the ossicular chain is rigidly fixed (with negligible inertia) and a second branch in series with the ossicular chain, corresponding to the portion of the eardrum that may be considered rigidly coupled to the malleus (and incorporating the effect of the middle-ear air cavity). Zwislocki (1962) refined his model by adding an inductor to represent tympanic membrane mass. An electric analog of the middle ear without incus and the otosclerotic ear was also analyzed (Zwislocki, 1963). Møller (1965) modeled the cat and rabbit tympanic membrane as a single branch in parallel with the branch representing the ossicular chain and cochlea, with the part of the tympanic membrane tightly coupled to the malleus being implicitly included in the latter branch. Peake and Guinan (1967), in their model for the cat, did not include a parallel branch for the tympanic membrane.

In the early 1970s, tympanic membrane vibration-pattern measurements by laser holography made it clear that the mode of tympanic membrane vibration described by von Békésy was incorrect (Khanna and Tonndorf, 1972). Even at low frequencies, no part of the drum acts like a rigid plate. Thus, Shaw and Stinson (1983, 1986) reinterpreted the nature of the two tympanic membrane "piston" components, associating them with (1) the small part rigidly coupled to the manubrium and (2) all of the rest of the tympanic membrane. Kringlebotn (1988) did not include multiple branches for the tympanic membrane itself but did include a parallel branch after the series branch, to represent coupling between the tympanic membrane and the manubrium. In an attempt to deal with higher frequencies, Puria and Allen (1998) presented a delayline model of the tympanic membrane: it was further explored by O'Connor and Puria (2008) and extended by Parent and Allen (2007, 2010). Teoh et al. (1997) investigated the effects of the large pars flaccida of the gerbil tympanic membrane. Two-port models (Shera and Zweig, 1991; O'Connor and Puria, 2008) were particularly appropriate for describing experimental data of tympanic membrane behavior by a single number such as umbo displacement or, especially, acoustical impedance. However, it is clear that lumped models cannot model the spatial vibration patterns of the tympanic membrane, nor can they address questions arising from those patterns. One disadvantage of circuit models is that their parameter values may need to be changed to accommodate changes in modes of vibration because of frequency, muscle contractions, different applied loads, or other effects. This greatly limits their predictive power. As is the case for the tympanic membrane, however, both circuit models (O'Connor and Puria, 2008) and two-port models (Shera and Zweig, 1992a, 1992b) of the ossicular chain continue to have value as concise representations of experimentally observed phenomena.

#### 3.2. Tympanic membrane and middle ear numerical models

Starting from the first FE model of the cat tympanic membrane implemented by Funnell and Laszlo (1978) several FE models have been used to investigate the static or dynamic behavior of middleear subsets or the entire middle-ear segment under normal or pathologic conditions. Ladak and Funnell (1996) modified a previous model of the cat tympanic membrane (Funnell et al., 1987) by adding explicit representations of the ossicles and cochlear load. In the 1996 model, the tympanic membrane was modeled as a curved conical shell, used to investigate sound transmission in the normal cat middle-ear for frequencies below 1 kHz and then modified to simulate the effects of two types of middle-ear surgeries. In Wada et al. (1992) a human middle-ear FE model including tympanic membrane, ossicles with fixed rotational axis, and cochlear impedance was implemented for the first time. The validity of the model was later extended (Wada et al., 1996) by including anterior mallear and posterior incudal ligaments, tensor tympani and posterior stapedial tendons, middle-ear cavity (as a rectangular solid), and ear canal (as a rigid tube) and the frequency response to an acoustic excitation was derived for frequencies below 3 kHz. The model was then further modified by adding the ligaments, tendons, incudostapedial joint, external ear canal, and middle-ear cavities (Koike et al., 2002). The FE model described in Williams et al. (1996) included the tympanic membrane, ossicles, and outer ear canal. The anisotropic property of the tympanic membrane was included in this model by introducing pseudo-fibers and diseases of the middle-ear were investigated by changing the model parameters. Hyperelastic Mooney-Rivling material properties and geometry nonlinearities were introduced by Wang et al. (2007) in their 3D model of human middle-ear in order to investigate its dynamic behavior. The analysis shows that the reduction of the tympanic membrane and footplate vibration magnitudes under positive middle-ear pressure are mainly determined by stress dependence of elastic modulus. On the other hand, the reduction of the tympanic membrane and footplate vibration magnitudes under negative pressure was caused by both geometry changes of middle-ear structures and the stress dependence of elastic modulus. In 2008, 3D models of the middle-ear cavities and portion of the ear canal from microCT imaging were introduced by Tuck-Lee et al. The effect of perforation and repair with patches was also explored. In the same year, Homma et al. (2009a) showed the presence of two ossicular modes below 2 kHz. Their numerical model revealed that the dominant mode under air conducted excitation is the first mode and is characterized by hinging ossicular motion whereas the dominant mode under bone conducted excitation is the second mode and is characterized by pivoting ossicular motion. Edom et al. (2013) studied the effect of rocking versus piston-like stapes motion on the inner ear hydrodynamics. They showed that the piston like motion evokes higher amplitudes than the rocking one and that this difference increases with increasing frequency. In addition the piston-like stimulation excites the basilar membrane along its entire length while the rocking motion is able to determine the vibration just of the portion of the basilar membrane directly below stapes. The coupling effect between ossicular chain and the transducer of implantable middle-ear hearing devices and the effect of different stapes prostheses on the round window (RW) vibration and the basilar membrane displacement were investigated respectively in Wang et al. (2011) and Kwacz et al. (2013, 2014). The main findings were that the piston stapes prosthesis causes an approximately fivefold lower amplitude round window vibration compared with normal anatomical conditions, a firm crimping connection between the clip and the ossicles can maximize the efficiency of the transducer stimulation. More recently Maftoon et al. (2015) provided evidence of the strong effect of Young modulus, thickness, and density of the pars tensa, Young modulus of the stapedial malleus annulus ligament and Young modulus and density of the mallus on the response of a FE gerbil middle ear model. Then, Gan et al. (2016) and Wang and Gan (2016) used a 3D FE model of the chinchilla TM and middle-ear to compute the distribution of stress in the TM and the TM displacement with impulse pressure waves (Fig. 1).

#### 4. Inner ear models

#### 4.1. Inner ear physical models

Physical models of the cochlea have been developed to investigate aspects of hearing function that are difficult to study in vivo due to the inaccessibility of the cochlea. Historically, we can distinguish two periods on the development of physical models of the cochlea. Because of the difficulties in fabricating and driving



**Fig. 1.** Finite element model of chinchilla tympanic membrane (A, B). Distributions of the equivalent von Mises stress (C) and the displacement (D) in the open case and the distributions of the equivalent von Mises stress (E) and the displacement (F) in the shielded case. All the results are relative to the time when the maximum stress was reached (readapted from Gan et al., 2016).

such a small device, the early mechanical cochleae were typically one or two orders of magnitude larger than a human cochlea. However, the material properties of these artificial cochleae were usually not scaled properly. For instance, Von Békésy (1960) built a mechanical cochlea to study the traveling wave. Tonndorf (1959) adopted a similar structure to study beats in the cochlea. Chadwick and Adler (1975) constructed a mechanical cochlear model to examine their theory of the cochlea function. Much later, Lechner (1993) implemented a cochlea with a Polyvinylidene Fluoride (PVF<sub>2</sub> or PVDF) bending transducer, which was mounted on beams on the artificial basilar membrane (ABM) to obtain an electrical output from the vibrational activity. Lechner also added actuators to a beam to simulate the active response in the cochlea. About the same time, Zhou et al. (1993), proposed the first physical model with life-sized dimensions for the basilar membrane. However, the basilar membrane thickness was not controlled and a fluid viscosity 20 times larger than the perilymph viscosity was used.

The second period began in the mid-1990s with the

introduction of improved micromachining technology. Haronian and MacDonald (1996) proposed a micro-electro-mechanically based frequency signature sensor (MEM-FSS). The model comprised an array of beams of gradually varying lengths of silicon beams lie on a core backbone of the same material. This core backbone was used to transfer vibrations along the device, simulating the fluid channel's function. In 2000, Manoussaki and Chadwick used a straight cochlea model with fluid loading and a more realistic coiled cochlea model to investigate the influence of geometry on the coupling between fluid and cochlear partition (CP). By comparing the fluid loading in the two cases, the pressure on the BM was found to increase with enlarging BM width and decrease along the cross-sectional area of the cochlear ducts. The same authors proved that: (1) coiling reduces the inertial impedance to fluid motion induced by the traveling wave; (2) in a coiled model, the pressure on the BM decreases as the pitch (and thereby the radius of curvature) of the cochlear coil decreases; (3) in a cochlea of realistic geometry, impedance reduction is greatest at the apex, the low-frequency end of the cochlea compensating for the great curvature. Lim et al. (2000) fabricated a mechanical cochlea by laying an isotropic film over a slotted silicon wafer and surrounding it with fluid channels. They demonstrated the propagation of the traveling wave by showing the phase plot at two locations along their model; however, the magnitude response shown is rather irregular and does not show the expected highfrequency roll off. Hemmert et al. (2003) proposed a fluid-filled Micro-Electro-Mechanical Systems (MEMS) based mechanical cochlea. In this model, the BM was built using an epoxy-based photoresist, which possesses stiffness close to the biological material. The authors used an impulse response at two very closely spaced locations to demonstrate the existence of a traveling wave.

More recently, Wittbrodt et al. (2004) proposed a new design continuing Lim's effort (Lim et al., 2000). Aluminum fibers were deposited on a soft membrane to produce the anisotropic artificial basilar membrane (ABM). In 2002, White and Grosh (2005) reported their cochlea MEMS, which used a single fluid channel and a BM made of silicon beams. White and Grosh sought to improve their previous model introducing silicon-nitride beams to reduce the residual tension on the beams (White and Grosh, 2005). Both the silicon and silicon-nitride beams are, however, much stiffer than their biological counterpart and this mismatch in material properties and anisotropy was assumed as the cause of abnormalities in the magnitude response of the beam responses. More recently, Jang et al. (2015) proposed a MEMS-based piezoelectric artificial basilar membrane (ABM) composed of a micro-electromechanical system cantilever array. The fabricated ABM was assembled incorporating a signal processor and an intra-cochlear electrode array. When applying acoustic sounds to the ABM, a cantilever in the ABM generates a piezoelectric output, which is converted by the signal processor into a stimulating electrical signal. The output of the signal processor was linked to the intracochlear electrode array inserted into the cochlea of a deafened guinea pig. Electrical stimulus was used to elicit evoked auditory brain responses (EABRs) from the auditory neurons in the guinea pig. Although Jang et al.'s ABM envisions the development of a lowcost cochlear implant containing an ABM, the dimensions of the signal processor incorporated and energy supply issues made the direct implantation impossible. Even nowadays, with the most recent innovation in microfabrication, the accuracy of physical models in mimicking the mechanics of the cochlea is surprisingly low if compared with other modeling strategies, such as numerical models, which can easily take advantage of improvements in computational power.

## 4.2. Inner ear analytical models

## 4.2.1. Analytical models of cochlea hydrodynamics

A direct consequence of the experimental work of Von Békésy on the mechanisms of hearing in the cochlea was the development of analytical models to explain the cochlea hydrodynamics and the mechanics of vibration of the basilar membrane. Most early models (Ranke, 1950; Zwislocki, 1948) were based on a 3D cochlear structure comprising 2 fluid-saturated chambers, the scala vestibule and the scala tympani interconnected at the helicotrema. It was considered that the fluid moves in such 3D chambers due to the one-dimensional displacement of the stapes. In the model by Zwislocki (1948), the fluid motion inside the chamber leads to an acoustic longitudinal or plane wave with a wavelength larger than the dimensions of the cochlea cross section. The analytical model by Peterson and Bogert (1950) was developed without the assumption of fluid incompressibility. They also discussed the role of the fluid viscosity, whose effect was also studied by Zwislocki (1948, 1950, 1953) and Fletcher (1951). Dallos (1973) developed a general equation for the Fourier transform of the pressure that contained two extra terms, the first as a fluid compressibility correction term, and a second for correction of viscosity effects. Peterson and Bogert (1950) followed the same approximation for a plain wave within the ducts, but they hypothesized that the speed of the wave decreases significantly at and beyond the point of resonance, resulting in a proportional decrease of the wavelength. In turn, Ranke considered the whole duct as a shallow water case, in which the water surface is covered by the cochlear partition that has very different properties than the fluid. However, the complexity of the CP is not comparable with a water-air interface and the cochlear partition impedance is very different to the water-air surface tension, since the properties of the cochlear partition depend on the location along the longitudinal direction, which is not the case for water-air interfaces. Also, analysis of surface waves requires the effect of gravity, but for the cochlear partition, no such effect is considered. Similar approaches were adopted by Siebert (1974), Lighthill (1981) and others.

More recent models of the mechanical response of the cochlea involve the forward propagation of a single "slow wave" (Zweig et al., 1976; de Boer and Viergever, 1984). This wave is generated by an interaction between the inertia of the fluid in the chambers of the cochlea and the stiffness of the BM and can be reproduced using simple one-dimensional box models (de Boer, 1996). At low sound pressure levels the amplitude of this wave is amplified by a number of active processes within the OC, but the basic description of slow wave propagation is valid even when the cochlea is passive and for high sound pressure levels. Since the properties of the cochlea, particularly the BM stiffness, vary along its length, the properties of this slow wave are position-dependent and frequency dependent. These properties can be characterized at each position along the cochlea by a complex wavenumber; the real part determines the wave speed and the imaginary part determines the spatial attenuation of the wave. Duifhuis (2012) derived equations for the pressure and for the velocity across the CP, which are equivalent to the models in Zwislocki (1953) and de Boer and Mckay (1980). These equations describe the behavior of the cochlear partition at different positions along the cochlea. Since the impedance of the cochlear partition is modeled as a second order system, the objective of describing the tonotopic response of the cochlea was readily achieved. Local tuning depends on local resonance properties, in other words, the ratio between local stiffness and mass, not their specific magnitude. Varin and Petrov (2009) and Reichenbach and Hudspeth (2014) used Navier–Stokes equation to describe the perilymph hydrodynamics and the surface wave that rise at the interface between BM soft tissue and endolymph. Assuming the acoustic impedance, Z, is constant, two solutions and, hence, two modes of wave propagation exist in the cochlea. In the first solution, the pressures are the same on both sides of the membrane. The membrane's displacement thus vanishes and the wave emerges from longitudinal variation only. In the second solution, the pressure changes on both sides of the membrane are equal in magnitude, but have opposite sign. In this case the membrane displacement does not die out. Regarding the wavelength, two limiting cases are important (Lighthill, 1981). First, when the wavelength is much smaller than the height of the channels, the fluid velocity and pressure decay exponentially with the vertical distance from the membrane. Fluid particles undergo circular motion, which for a forward travelling wave (TW) is clockwise below and anti-clockwise above the BM. The motion near the cochlear walls is essentially zero, so those boundaries have no influence on the wave. In the second limiting case, the wavelength greatly exceeds the height of the chambers: fluid particles move on elliptical trajectories, for which motion occurs predominantly in the longitudinal direction. Again, the longitudinal component of velocity barely depends on the vertical position (Fig. 2).



**Fig. 2.** Simplified cartoons representing: (A) structure of the ear; (B) longitudinal projection of the cochlea in its uncoiled form; (C) cochlear transverse section and interior and interior structures; (D) 2-D model of the uncoiled cochlea with SV and SM united in one single chamber; (E) surface view of the basilar membrane with a schematic representation of the collagen fibers running from the neural to the abneural edge; wave propagation in the cochlea (F) for small wavelength,  $\lambda \ll h$ . In this case the trajectories of liquid particles are circular and their amplitudes decay exponentially away from the BM. Each arrow's head denotes the position of a particle at a given time; (G) when  $\lambda \gg h$  and the motion occurs predominantly in the longitudinal direction (readapted from Reichenbach and Hudspeth, 2014).

#### 4.2.2. Analytical models of basilar membrane vibration

As described in the previous section, the complexity of the wave equation for the cochlear partition in the frequency domain raises from the spatial variability of the square of the wave number,  $k^2$ . An analytical solution was devised for the case in which  $k^2$  varies slowly with x, which is named after its developers, Wentzel, Kramers, Brillouin and Jeffreys, or WKBJ (also referred to as WKB) method. The WKBJ approach was first applied to the mechanics of the cochlear partition by Zweig et al. (1976) and then de Boer (1980). The mechanical response of the cochlea can be calculated using the WKBJ method based on the wavenumber distribution along the cochlea obtained using an inverse method (Shera, 2007). The WKBJ method has a number of inherent assumptions, such as that the wave is only traveling in one direction. This implies that no backward traveling wave is generated by the normal hearing function of the cochlea. Another assumption is that the wavenumber does not vary too rapidly with position, as compared with the wavelength (de Boer and Viergever, 1984), although this assumption appears to limit the applicability of the WKBJ method in cochlear modeling less than one would expect (de Boer, 1980). Zwislocki (1948, 1974) predicted the delay of the traveling wave to accumulate with increasing distance from the stapes. Steele (1974) firstly adopted the WKBJ method to solve cochlear mechanical problems and found closed-form solutions for a 1D cochlear model. Zweig et al. (1976) found the closed-form WKBJ solutions for a 1D long-wave model in 1976. Steele and coworkers also extended the WKBJ method to solve 2D (Steele and Miller, 1980) and 3D (Steele and Taber, 1979a,b; Taber and Steele, 1981) cochlear problems. The 3D model developed by Steele and Taber (1979b) of the WKBJ approximation indicate that both the tapering of the scalae and the damping have little effect on the slow wave velocity. de Boer and Viergever (1982, 1984) further developed the WKBJ approach for cochlear mechanics. The WKBJ solutions for the 2D and 3D cochlear model showed good agreement with more detailed numerical solutions, except for the region just beyond the BM response peak, which was suggested to be due to the non-uniqueness of the complex WKBJ wavenumber in 2D and 3D models (Ranke, 1950; Lesser and Berkley, 1972; Lien, 1973; Viergever, 1977; Allen, 1977; Allen and Sohndi, 1979; Steele and Taber, 1979a; de Boer, 1979; Holmes and Cole, 1984; Chadwick, 1985; Diependaal and Viergever, 1989).

## 4.3. Inner ear numerical models

Few FEA models have so far investigated the hydrodynamics of the inner ear. Kassademi et al. (2005) reported the endolymphatic velocity and pressure fields on the lateral semicircular canal system of the inner ear. Edom et al. (2013) studied the effect of a piston like or rocking stapes motion on the cochlear fluid flow and the BM motion in a 2D box model of the cochlea. In De Paolis et al. (2016) a detailed 3D microCT-based human cochlea model of the perilymph hydrodynamics was investigated. Elliott et al. (2013) decomposed the full BM responses of both passive and active cochlear models in terms of wave components. They found that besides the conventional slow wave, an evanescent, higher-order fluid wave starts to make a significant contribution to the BM response in the region apical to the peak location. Zhang and Gan more recently developed a more realistic model with a coiled cochlea coupled with the middle-ear and ear canal, with viscoelastic material properties implemented for the middle-ear soft tissue (Zhang and Gan, 2011, 2013) (Fig. 3C). The authors were able to compute the middle-ear transfer function and the BM vibration displacement normalized



Fig. 3. Evolution of inner ear finite element models (reproduced with permission from (A) Bohnke and Arnold, 1999; (B) Gan et al., 2007; (C) Zhang and Gan, 2011.

by the stapes displacement as a function of the distance from base. The model was also used to compute the efficiency of the forward and reverse mechanical driving with middle-ear implant, as well as the passive vibration of the BM when a cochlear implant was placed in the ST. Compared with the simplified 2-chamber straight cochlea model, the 3-chamber spiral cochlea model provided more realist geometric and physical information. Interesting, the new model also predicted that the material properties of Reissner's membrane (RM) affected the cochlea function by influencing the vibration of the BM.

One of the first models to accurately simulate BM vibration characteristics was developed by Bohnke and Arnold (1999). It was a 2-chamber 3D finite element passive model constructed using microCT (Fig. 3A). The perilymph was considered inviscid and the input value was a pressure load of 1 Pa at 3 frequencies (100 Hz, 2000 Hz, and 10000 Hz). The cochlea partition, on the other end, was represented by shell elements with constant Young's moduli along the radial and longitudinal direction and damping coefficient. The predictions of the BM displacement were in agreement with observations by von Békésy, even though the authors noted that the damping was the only parameter responsible for the phase shift of the BM displacement. Gan et al. (2007) developed a passive. uncoiled 2-chamber, fluid filled FE model of the cochlea, coupled to a model of the ear canal and the middle-ear (Gan et al., 2004, 2005). The BM is represented as a 32 mm trapezoidal surface whose dimensions are assumed to change linearly in width from 0.1 mm at the base to 0.5 mm at the apex and in thickness from 7.5  $\mu$ m at the base to 2.5  $\mu$ m at the apex (Fig. 3B). The BM boundaries are fully restrained (all displacements and rotations zero) at the spiral lamina and simply supported (all displacement zero) at the spiral ligament, consistently with the anatomical studies by lurato (1962). After Gan's first attempt (2007), several 2-compartment uncoiled

models of the cochlea were implemented with or without the middle/external ear components. In all these models the perilymph and, when present, the endolymph is generally modeled as a viscous fluid, similar to water with a bulk modulus of approximately 2.2 GPa. The BM is generally assumed to have a length of 32–35 mm, a Poisson's ratio equal to 0.3 and a mass density of 1200 kg/m<sup>3</sup>. Liu and White (2008) showed that the value of the Poisson's ratio does not have a major impact on the vertical displacement of the basilar membrane.

## 5. Micromechanical models of the organ of Corti

With the development of more refined measurement technologies, more and more evidence showed that the cochlea is active and nonlinear. Recent progress on Optical Coherence Tomography (OCT) (Chen et al., 2011: Zha et al., 2012: Lee et al., 2015) has even allowed the non-invasive visualization of the TeM and BM in vivo (Lee et al., 2015). The basic mechanism of the active process in the in vivo cochlea can be explained as an interaction between the BM and the outer hair cells (OHCs). When the BM moves upwards, the stereocilia of the OHCs are deflected by the shearing motion between the reticular lamina (RL) and the TeM, which opens ion channels and causes a change in the OHCs intracellular potential and thus a change in the length of the cell, which will generate a force upward acting on the BM from the OHCs through the Deiters cells to enhance BM motion. The first finding related to the nonlinearity in the cochlea was back in 1971, when Rhode (1971) pointed out that the BM response is less frequency selective for higher-level sinusoidal stimulation. The idea of active processes in the cochlea was first raised by Gold (1948) and Kemp (1978) in the form of objective tinnitus and otoacoustic emissions. These active processes provide a frequency sharpening mechanism. Evidences



**Fig. 4.** (A) Strelioff's network model (1973). In this model, the cochlea is considered in cross-section slices. Each slice has six transverse resistors ( $R_1$  to  $R_6$ ) and five longitudinal resistors ( $R_7$  to  $R_{11}$ ) (reprinted with permission from Strelioff, 1973). (B) The electrical network model of the organ of Corti with dependent current sources instead of variable resistors at a given cross-section (reprinted with permission from Ramamoorthy et al., 2007). The SV, SM, and ST cables run longitudinally along the cochlea.  $\phi_{5T}^+$  represents ST potential at an apical location ( $x + \delta x$ ) in the cochlea relative to the other quantities. Is and Is are the dependent current source for apical and basal surfaces, respectively.  $\frac{1}{3}R_a^0$  and  $3C_a$  model the apical resistance and capacitance.  $\frac{1}{3}R_m$  and  $3C_m$  model basolateral resistance and capacitance. The longitudinal electrical coupling is similar to Strelioff (1973), as shown in Fig. 4A. (C) An electrical model of the inner hair cell (reproduced with permission from Lopez-Poveda and Eustaquio-Martin, 2006). In this model, the apical conductance  $w_A$  is a function of the IHC stereocilia displacement, u(t).  $C_A$  and  $C_B$  model the apical and basaliar membrane of the IHC.  $R_r$  and  $R_p$  are epithelium resistances.  $V_M$  is membrane potential.  $w_{K,f}$  and slow resistances.

of the active behavior in the living cochlea are the higher gain at lower stimulus level and the stimulus level dependence of the tuning. Other evidence of the active behavior in the living cochlea is given by the detection of sound in the ear canal, due to spontaneous oscillations originating from the cochlea, retransmitted by the middle ear, in the absence of any excitation (Wilson, 1980).

## 5.1. Lumped parameter models of the organ of Corti

Current models of the micromechanics of the OC often use a lumped parameter representation of the BM, TM, and the structures into which the hair cells (HC) are embedded. To achieve a better understanding of cochlear function, cochlear micromechanics models should be able to explain how the vibrations of the cellular and membranous components of the cochlear partition result in deflections of the inner hair cells (IHC) stereocilia. A lumped-parameter model by Davis (1965) is an initial attempt to model the electrical network properties and the distribution of potentials in the cochlea. In this model, the cochlear resting potentials have been modeled by two batteries: a primary battery in the hair cells and a secondary battery in the stria vascularis, and the mechanical electrical transduction (MET) channels are modeled by variable electrical resistors. Accordingly, the current through the hair cells is modulated by changing electrical resistances resulting from cilia deflection. Strelioff (1973) suggested a network model of the resistors and batteries to simulate the generation and distribution of the cochlear potentials, as shown in Fig. 4A. The results of this model were in agreement with previous physiological findings. In Dallos (1983, 1984) electrical properties of the organ of Corti have been investigated. Allen (1980) derived the relationship

between the transverse motion of the BM and the shearing motion experienced by the OHC stereocilia. In his model, the TM is assumed only to rotate with the same angular movement as the BM. If the TM is allowed to move radially, the OC can be expressed by a two-degree-of-freedom model, in which the BM and the TM are assumed to move only in a single direction. It is also possible to apply the active force generated by the OHC on the model, as suggested by Neely and Kim (1986). An alternative active model is one in which the force is assumed to act across a very stiff OC, resulting in an active displacement, as in the model of Neely (1993). More detailed lumped-parameter micromechanical models have been proposed. Lim and Steele (2003) adopted a hybrid WKBJ numerical solution for their nonlinear active cochlear model, in which the lumped-parameter model of the cochlea is a simplification of the OC. In this kind of model, the properties of the spatially distributed OC are represented by a topology consisting of discrete entities (masses, dampers and springs) that approximate the dynamic behavior of the OC under certain assumptions. Mechanically, every component in the lumped-parameter model is taken as a rigid body and the connection between each rigid body takes place via springs and dampers. The model can be divided into a finite number of segments in the longitudinal direction with each individual segment having a unique characteristic resonant frequency, decreasing from 20 kHz, at the base, in the human, to about 200 Hz at the apex over the 35 mm BM length.

In Ramamoorthy et al. (2007), a model has been proposed which integrates the electrical, mechanical, and acoustical elements of the cochlea and provide an equivalent configuration for the imbedded OHCs (Fig. 4B). This model provides a framework to successfully predict and reproduce the response of the cochlea to acoustical stimulus comparable to experimental data. Nonlinear characteristic of the MET channel and hair bundle motility have not been considered in this model. A notable observation from this model is that longitudinal electrical coupling actually sharpens the mechanical response. This effect is also reported in Meaud and Grosh (2011) and is in part the motivation for the models of Iwasa and Sul (2008) and Dimitriadis and Chadwick (1999). Whether this effect is significant in a nonlinear cochlea is not known. Nonlinear saturation behavior of the MET channel has been incorporated in Liu and Neely (2010) to explore distortion product otoacoustic emission. In this model the longitudinal electrical connection in the organ of Corti and the hair bundle motility have been neglected. The model of Nam and Fettiplace (2010, 2012) has used a mechanical model along with electrical coupling to investigate the effects of the hair bundle motility and the cochlear amplifier in high and low auditory frequencies. Electrical properties of hair cell in vitro and vivo have been examined and very sophisticated models with detailed ion channels are reported in the literature (O'Beine and Patuzzi, 2007). This model or slightly different versions of it have been widely used to model the distribution of cochlear biopotentials and their effects (Dallos, 1983, 1984; Dimitriadis and Chadwick, 1999; Iwasa and Sul, 2008; Mistrick et al., 2009; Cheatham et al., 2011; Johnson et al., 2011). Fig. 4C illustrates a detailed circuit model of the IHC. This model is used to investigate the contribution of IHC to auditory compression (Lopez-Poveda and Eustaquio-Martin, 2006).

#### 5.2. Numerical models of the organ of Corti

Kolston and Ashmore (1996) built a 3D network cochlear model using 3D finite elements with individual cellular and membrane components of the OC being embedded within the fluid. The results from their simulations suggest that both the TM radial stiffness and especially the Deiters cell axial stiffness play a crucial role in the OHC-BM feedback loop. Similarly, Nam and Fettiplace (2010) and Nam (2014) developed an elastic micromechanical model of the organ of Corti to study force transmission and elastic wave propagation in which OHC somatic and hair bundle active forces were both considered. Fluid-solid interaction FE models (Cai and Chadwick, 2003, 2004) allowed studying detailed motion patterns within the OC in response to acoustic excitation (Steele et al., 2009; Yoon et al., 2007; Yang et al., 2013; Liu et al., 2015; Baumgart et al., 2008; Ni et al., 2016). Cai and Chadwick (2003) developed a hybrid approach based on the WKBI approximation to model the cross-sections of the guinea pig cochlea at several positions along the cochlea and solved the fluid-solid interaction eigenvalue problem for the axial wavenumber, fluid pressure, and vibratory relative motions of the cochlear partition as a function of frequency. Andoh and Wada (2004) predicted the characteristics of two types of cochlear pressure waves, fast and slow waves, and later estimated the phase of the neural excitation relative to the BM motion at the basal turn of the gerbil, including the fluid-structure interaction with the lymph fluid (Andoh et al., 2005). A two-dimensional finite element model of the OC was constructed based on measurement in the hemicochlea of the gerbil (Edge et al., 1998). The fluid within the Corti tunnel was treated as an elastic body without shear stiffness and the viscous force was considered analytically on the assumption that Couette flow occurs in this space. The effect of the mass of the fluid in the subtectorial space was assumed to be negligible. The SV and the ST were constructed in a 3D form to simulate the behavior of the lymph fluid and its interaction with the OC. The dynamic behavior of the local section of the OC, which extends in the longitudinal direction, was simulated and longitudinal widths of both fluid models were determined to be 48 um. which was less than one-fourth of the wavelength of the traveling wave (Ren, 2002; Ren and Nuttal, 2001). Indeed more recent studies by Ren et al. (2011, 2015, 2016) showed experimentally in vivo the transverse and the longitudinal displacement of the cochlear partition and the role of the hair cell in driving the reticular lamina. In Zagadou et al. (2014) the experimental measurements of the OC's point-stiffness performed in Naidu and Mountain (2007) were simulated using a 3D linear elastic isotropic FE model to determine the elastic properties of each individual structural component within the OC in the middle turn section of the gerbil cochlea around the 4 kHz (Fig. 5). The geometry of the model was based on direct measurements of anatomical feature within the OC. The distribution displays the material heterogeneity within the OC associated with the spatial organization of the cellular components. The mean value of the Young's modulus ranges from 2.41 kPa to 10 GPa for the cochlear section studied. Specifically a value of 3 GPa was found for the stiffness of the pillar cells (PC).

## 6. Whole hearing system models

#### 6.1. Whole hearing system analytical models

Research involving models of the entire human hearing system are essential to understand sound transmission mechanisms, to evaluate hearing loss treatments, to develop implantable devices and prostheses or to simulate pathologic conditions. Few wholehearing system models have been developed to date. A onedimensional vibration network model was successful in predicting empirical data measured on human temporal bone (Kringlebotn, 1988). More recently, Feng and Gan (2004) proposed a lumped parametric model of the human ear consisting of masses suspended by dashpots and springs representing the air inside the external canal, the tympanic membrane and the three middle-ear ossicles. Five model parameters, namely TM annulus stiffness, incus-stapes joint stiffness, stapedial annulus stiffness and middleear ligaments, exhibited a significant effect on the dynamic behavior of the model. A much more comprehensive hearing



**Fig. 5.** Cochlea cross section implemented in the FE model. (A) Histological section of the middle turn gerbil cochlear cross section. (B) FE representation showing the parts of the cochlear cross section. Tectorial membrane, TM; OSL, osseous spiral lamina; Ph, phalangeal supporting cells; HC, hair cells; LIG, spiral ligament. (C) Static point-stiffness experiments to determine the Young's moduli. Shown are the deformations of the structures in response to a probe forced under BM at a particular radial location. (Picture adapted from Zagadou et al., 2014).



Fig. 6. FE models of the inner ear. The ossicles are coupled with the cochlea which is represented with: (A) a straight model in Kim et al. (2011); (B) a coiled geometrical model in Zhang and Gan (2011) (reproduced with permission from Kim et al., 2011; Zhang and Gan, 2011).

system model was developed by the Army 25 years ago, which is described in more detail next.

## 6.2. Whole hearing system numerical models

Finite element models of the human ear have greatly evolved in recent years based on microCT images. In 2004, Gan et al. used a 3D FE model of the human ear comprising detailed morphology of external ear canal, middle-ear cavity and middle-ear ossicles to study the effect of tympanic membrane thickness and stiffness, incudostapedial joint material, and cochlear load on acousticmechanical transmission through the human ossicle chain (Gan et al, 2004). The same model was utilized to investigate the influence of tympanic membrane perforations (Gan et al., 2005) and otitis media (Gan et al., 2007) on the pressure distribution in the external canal and middle-ear cavity. The results showed that variation of middle-ear pressure is related to the perforation type and location and the frequency. This model was further improved with the introduction of an uncoiled cochlear model with two straight fluid channels separated by the basilar membrane with orthotropic material properties and dimensional variation along its length (Gan et al., 2007 and Gan et al., 2009). The model was used for studying the middle-ear transfer function in normal and pathologic conditions, sound pressure gain across the middle-ear and the cochlear input impedance in response to sound stimulation. The analyses performed in Kim et al. (2011) allowed additional clarification of the basilar membrane vibrational pattern in presence of bone conducted (BC) and air conducted (AC) stimulations (Fig. 6A). The invariant basilar membrane characteristics for BC and AC excitations confirmed that the basilar membrane is only driven by the component of the fluid pressure that is antisymmetric (i.e. differential slow wave) and that the basilar membrane only responds to the excitation generated at the two windows regardless of whether these are produced by BC or AC. In Wang et al. (2014) an auditory periphery model integrating ear canal, middle-ear and active cochlea was provided as a tool to explore the mechanism by which sound pressure in the ear canal is converted to a stimulus for the outer hair cells. Finally in the human ear model implemented in Zhang and Gan (2011, 2013) a spiral cochlea model with three scalae, SV, ST and scala media (SM), was introduced for the first time (Fig. 6B). Viscoelastic material properties were also applied to the middle-ear soft tissue and the energy absorption in the normal ear and in presence of three middle-ear disorders: otitis media, otosclerosis and ossicular chain disarticulation. The model also predicted that the material properties of the Reissner's membrane (RM) affect the cochlear function by influencing the vibration of the

## 7. Hearing damage

Hearing damage through blast is an escalating problem in the military: it accounted for 25% of all injuries during Operation Iraqi Freedom in 2004 and was accordingly the most common single injury (Gondusky and Reiter, 2005; Mathews and Kovfman, 2015). Auditory dysfunction in general is now the most prevalent individual service-connected disability, with compensation estimated to exceed \$2.2 billion in 2014 in the U.S.A (Fausti et al., 2009). Injuries resulting from blasts are categorized into four groups (Choi, 2012: Ritenour and Baskin, 2008). Primary blast injuries, in particular, are caused by over- or under pressurization from the blast and most commonly affect air-filled and fluid-encased organs within the body. Otologic injury, typically considered a primary blast injury, can result in reduced hearing and can compromise a warfighter's situational awareness immediately and permanently and it can hamper a service member's ability to meet mission requirements, remain in his or her military occupational specialty, and maintain fitness for duty.

Blast overpressure is a high intensity disturbance in the ambient air pressure that creates high intensity sound (impulse) over 170 dB SPL (Gan et al., 2016). The over-pressurization of air molecules and the resultant impulse noise (characterized by excessive peak pressure levels) can cause temporary or permanent auditory threshold shift (PTS), or hearing loss. The tympanic membrane rupture occurs in 50% of adult ears after a blast exposure (Chandler and Edmond, 1997). Other middle ear damage, such as disarticulation of the ossicular chain or fracture of the ossicles, can also result from blast exposure (Chait et al., 1989). There is some evidence to suggest that TM perforation and ossicular disruption may give some protection to the cochlea. Akiyoshi et al. (1966) found less cochlear damage in guinea pigs exposed to a larger blast (resulting in disruption of the middle ear) than in those exposed to smaller blasts. Hamernik et al. (1984) and Eames et al. (1975) reported similar findings in animals exposed to multiple shock waves. More recently, Cho et al. (2013) estimated the impact of healed tympanic membrane perforations on auditory function by comparing the blast-exposed mice to mice that had surgically-perforated tympanic membranes and Luo et al. (2016) characterized the mechanical properties of human TM at high strain rates after multiple exposures to blast explosion. The results indicate that the blast wave have different effect on the mechanical properties in the radial and circumferential directions. Within the cochlea, the basilar membrane (BM) is the structure that is most vulnerable to the effects of acoustic trauma (Patterson and Hamernick, 1997). Acoustic overexposures causing moderate, but completely reversible, threshold elevation leave cochlear sensory cells intact, but cause acute loss of afferent nerve terminals and delayed degeneration of the cochlear nerve (Kujawa and Liberman, 2009). However, blast waves can tear the inner and outer hair cells away from their support cells. Electron microscopy has given considerable insight into the damage occurring within the cochlea. It is postulated that a temporary threshold shift results from changes in the integrity of the tight cell junctions of the reticular lamina, changes in the membrane permeability, or holes in the reticular lamina. These breaches allow mixing of the perilymph and endolymph, which alters the ionic environment and interferes with the physiological events within the cochlea. As the breaches in the reticular lamina are repaired the temporary threshold shift resolves. PTS in the chinchilla exposed to repeated 160 dB impulses have been associated (Hamernik et al., 1984) with damage that included loss of hair cells and even the rupture of long segments of the OC, which became detached to float in the SM alongside cellular debris. Cilia of the outer hair cells were found bent, fused or broken and some giant cilia were found damaged. The inner hair cells were damaged to a lesser extent. Holes were found in the reticular lamina with cytoplasmic extrusions at the site of maximal cochlear disruption and Claudius cells appeared to be able to spread their membranes to cover a considerable area of damage. Macrophages within the cochlea increased, reaching a peak at 12–16 days, though phagocytic activity was still evident at 30 days. Kumagami (1992) noted degenerative changes in the endolymphatic sac and some degree of endolymphatic hydrops in guinea pigs exposed to blast, though there are not usually any features of hydrops in human blast victims.

## 7.1. Numerical models

The role of the auricle and the external ear when exposed to high-pressure sound energy is of considerable interest for appropriate design of audiometric and hearing aid devices as well as the hearing protection devices. The effect of high pressure and blast waves on human external ear was investigated by Fadaei et al. (2014) using a computational fluid dynamics (CFD) study on a realistic model of the human external ear canal. The effect of the auricle, frequency and different peak pressure magnitudes of the blast wave on pressure distribution patterns within the ear canal were investigated. The results clearly demonstrated that the frequency plays a key role on pressure distribution within the ear canal. At a frequency of 4 kHz, the pressure magnitude at the TM is much larger than the pressure magnitudes corresponding to the frequencies of 2 and 6 kHz. The ear canal model with auricle exhibited 7% larger pressure magnitudes when compared to those of the ear canal model without the auricle tested for blast waves. In Homma et al. (2009b), a FE model of a human middle ear was used to gain further insight into the dynamics of the bone conducted and air conducted response alterations due to ear canal pressurization, which is a critical step toward the potential future utilization of this phenomenon for hearing protection in extremely noisy environments.

## 7.2. Nonlinear models

## 7.2.1. Nonlinear cochlea models

In a linear cochlear model, a sinusoidal pressure difference across the CP generates a sinusoidal BM velocity and so the relationship between them can be simply represented by an impedance function. In a nonlinear cochlear model, however, the waveform representing the BM velocity response to a sinusoid becomes distorted. In addition, while a linear system cannot generate signal components that were not present in the stimulus spectrum, any nonlinear system will produce harmonic distortion products in response to simple tonal stimuli, and more complex stimuli produce more complex distortion product spectra. The nonlinearity is typically obtained by defining nonlinear elements, such as nonlinear damping, nonlinear OHC force or pressure, and nonlinear geometry.

The relationship between the pressure difference, p, and BM acceleration,  $\ddot{w}$ , at different positions, x, along the cochlea can be expressed by the long-wave approximation as:

$$\frac{\partial^2 p(x)}{\partial x^2} = -\frac{2\rho}{h} \ddot{w}(x) \tag{1}$$

where the BM dynamics are related to the pressure difference as

$$p(x) = m(x)\ddot{w} + r(x,\dot{w})\dot{w} + s(x)w$$
(2)

in which m,  $r(x, \dot{w})$  and s(x) are respectively the mass, the damping and the stiffness of the system. The mass is usually assumed to be constant.

Kim et al. (1973) studied the nonlinear behavior of the cochlea using a 1D model, which is represented by a nonlinear differential equation. The damping term in this differential equation is a function of velocity and "effective frequency" which depends on the input level. They assumed the characteristic frequency to be the limiting value of the effective frequency as the input amplitude approaches zero. Using this model they predicted the distortion products and other nonlinear phenomena in BM motion. The model behaves effectively linearly at low input levels and nonlinearly at high input levels. In a similar way, Hall (1974) incorporated the nonlinearity in a transmission-line model of the cochlea by defining a nonlinear resistance and calculated the nonlinear BM response and distortion products in the spatial domain. In Van der Pol's oscillator theory (Duifhuis, 2012), the displacement of the oscillator is described by an ordinary differential equation similar to the one that describes a damped mass-spring system, but the damping term is a nonlinear term with a negative value for small amplitudes. On the contrary, for large amplitude, and also for large velocity, the damping term is set to be proportional to  $(\ddot{w})^2$ , which leads to a power of 1/3 dB/dB in the output-input level curve. Following the van der Pol oscillator model, Duifhuis proposed a modified Van der Pol oscillator model (Duifhuis, 2012) where the damping is a function of the hyperbolic sine of the velocity and is approximately constant for small amplitudes and increases exponentially at large amplitudes. The nonlinear model proposed by Furst and Goldstein (1982) was also very suited for time domain analysis. The function proposed by Furst and Goldstein contains a nonlinear damping profile, similar to the Van der Pol nonlinear profile, but lacking its characteristic near-zero negativity. An additional nonlinear correction was also applied to the stiffness term

#### 7.2.2. Force and geometrical nonlinearity of OHC

In addition to models describing nonlinear damping of the BM in the Cochlea, various force or pressure nonlinear models have been developed to describe the effect of the OHCs motility in the OC. Kanis and de Boer (1993) developed a 1D model to replicate the nonlinear behavior of the cochlea, containing a saturating pressure generator, located at the OHCs, which modifies the BM velocity via adding a nonlinear pressure to the pressure difference across the OC. The nonlinear pressure was defined using the hyperbolic tangent function and the model was solved with a "quasilinear" iterative method in frequency domain. Later Kanis and de Boer (1996) compared the result with that from the time domain solution proposed by Diependaal et al. (1987) and showed that the two methods match well for single-tone and two-tone stimulations. Chadwick (1998) treated the nonlinearity as a correction to a linear hydroelastic wave in his two-degree-of-freedom lumped-parameter model. The active force generated by the OHC was expressed as a function of the relative transverse displacement between the TeM and the RL and the saturation of this force was represented by the hyperbolic tangent function. Using a multiple scale approximation he was able to relate the active force function to the local wavenumber, which is a function of the BM displacement. In this way, the nonlinearity in the model depends not only on the TeM-RL displacement but also on the BM displacement. He also showed that the nonlinearity of the outer hair cell generates retrograde waves travelling backward toward the base.

Bohnke et al. (1999) constructed a complex 3D FE human

cochlear model including detailed OC structure to further analyze the active and nonlinear mechanics. In their model, only a section of the OC, including OHCs, was modeled. The load due to surrounding lymphatic fluid was represented by a symmetrical loading on the OC. The nonlinearity was represented by defining a nonlinear OHC function, which expresses the OHC receptor potential as a function of the stereocilia displacement, using a secondorder Boltzmann function. However, the effect of the nonlinearity was only roughly shown in terms of the distorted time signal. Bohnke and Arnold (1998) analyzed the geometrical nonlinearity caused by the phalangeal processes using a 3D FE human cochlear model including detailed elements within the OC, such as OHCs, RL, and Deiters cells. They considered large deflections, large rotations but small strains, and also the stiffening effect on the structure due to its state of stress. These two types of geometrical nonlinearities in the Deiters cells were modeled using a thin elastic beam with low modulus of elasticity. In their model, an external force due to two-tone stimulation is applied on top of the RL. This leads to a displacement of the OHC between the RL and the Deiters cells. When the model is linear, the amplitude spectrum of output displacement clearly shows two peaks corresponding to the two driving frequencies. When the geometrical nonlinearity is considered, however, distortion components appear in the spectrum.

Lim and Steele (2002) developed a 3D uncoiled feed-forward nonlinear cochlear model based on the WKBJ method. The model was built based on geometric and material properties from a chinchilla cochlea, in which the CP was modeled as an elastic orthotropic plate and viscosity of the cochlear fluid was also included. Fourier series expansions were used for an iterative procedure of solving the model. The model was used to predict BM response under different input levels, BM frequency response, BM velocity compression, distortion, and two-tone suppression. Comparison between numerical results and those from experiments shows a reasonable agreement. Following studies of OHCs mechanoelectrical transduction (MET), Liu and Neely (2010) construct a nonlinear model to study DPOAEs. The receptor current that flows into an OHC is defined as a nonlinear antisymmetric function of the RL displacement and velocity.

Comparing 2D and 3D nonlinear cochlear models, Diependaal concludes that the 2D response is close to the 3D solution but that the 1D solution deviates considerably from the multidimensional solutions (Diependaal, 1988). However, he points out that response from the 2D passive model with realistic values of the parameters may deviate significantly from the 3D response. So the dimensionality of a cochlear model depends on how much output one wants to obtain from the simulation. If one is only interested in cochlear macromechanics, a 1D approach is often satisfactory. However, if one wants information about interaction within the OC, cochlear micromechanics, a more detailed model, a 2D, sometimes a 3D, is required.

# 7.3. Auditory Hazard Assessment Algorithm for Human (AHAAH) model

The U.S. Army developed an Auditory Hazard Assessment Algorithm for Human (AHAAH) software (Price and Kalb, 1991; Price, 2006; 2007a, 2007b), based on a mathematical lumped model of the ear designed to predict auditory injury from blast exposures. The AHAAH model is a human full-ear circuit model comprising three components: the outer ear from pinna to tympanic membrane, the middle-ear including the air cavities and ossicles, and the inner ear with the cochlea. The three components are electroacoustically connected to each other to allow the propagation of external noise to the cochlea (Price and Kalb, 1991) which is modeled using WKBJ approximation. Briefly, it reproduces the measured transfer functions from the free field to the stapes and translates stapes motion into basilar membrane displacements. It keeps track of the displacements at 23 locations (roughly 1/3 octave intervals) and derives a dose at each location by squaring the peak amplitude of each upward displacement of the basilar membrane (in microns) and summing them for the analysis interval. The result is reported in auditory risk units (ARUs):

$$ARU = \sum d^2$$
(3)

where d = upward basilar membrane displacement (in microns).

Further, it has been empirically established in the cat ear that ARUs can be converted into compound threshold shift (CTS) at 30 min (Price, 2006):

$$CTS = 26.6 \cdot \ln(ARU) - 140.1 \tag{4}$$

where CTS includes both temporary and possible permanent components of threshold shift.

This function is presumed to be the generic relationship between ARUs and threshold shift for at least the cat and human cochleas; thus this formula can be used to predict specific amounts of CTS for any number of ARUs. Under intense exposures, where the loss mechanism is primarily mechanical stress, recovery begins only very slowly and is not likely to go to completion when a dose of 500 ARUs (25 dB CTS) is exceeded. When the 500 ARU limit is exceeded, the amount of permanent threshold shift (PTS) is usually about 0.6 of the loss at 30min (Price, 2006). Further, it has also been shown that the predicted PTS corresponds to hair cell loss (Price, 2006).

To determine its validity as a predictor of hazard for the human ear, data from the literature (a waveform and changes in hearing sensitivity) and from the US Army's "Albuquerque Studies" (53 different cases) were tested with the model. For comparison, MIL STD-1747(D) and A-weighted energy were also used to compute the hazard for the same data. The AHAAH model predicted correctly in 95% of the cases, the MIL STD-1474(D) was correct in 38% of the cases and A-weighted energy was correct in 25% of the cases. Errors for all methods tended to be in the direction of over-prediction of hazard. However, the AHAAH model showed a non-monotonic dose-response behavior against large weapon noise (Price, 2007b), which suggests that an exposure at higher intensity will result in lower risk of injury although this is not supported by field data (Smoorenburg, 2001). Other controversies around the AHAAH model regard the selection of the parameters, originally developed for the cat and then extended to the human model, the acoustic reflex (AR) modeling and the lack of a methodology to account for warned and unwarned exposures (AR) and the use of hearing protections.

To maintain a structure close to the ear's physiology, the model includes two non-linear elements present in the conductive path that are critical in predicting hearing loss. The first of these was the middle ear muscle response. Middle ear muscles act primarily by stiffening the middle ear (largely the effect of the stapedius muscle that rocks the stapes sideways and stiffens the ligament that holds it in the oval window). This results in attenuation of frequencies below the middle ear resonance most and the frequencies above it somewhat less (Pang and Peake, 1986; Wever and Bray, 1942).

Another form of non-linearity in the middle ear is that of the annular ligament of the stapes. The annular ligament is very rugged and simply cannot displace by more than 30 µm or so (Bolz and Lim, 1972; Guinan and Peake, 1967). Thus, when there are very intense impulses that would try to displace the stapes by 50 times that, the stapes becomes a strong peak-clipper (Price, 1974). The model includes this feature of the middle ear response. The presence of this clipping-characteristic of the ear has profound implications for rating hazard at very high intensities. For example, the presence of low-frequency energy in an impulse can produce displacements



Fig. 7. Schematic of MATLAB/Simulink AHAAH model showing the hearing organ block components including the outer, middle, and inner ear. The inset shows the details of the block for the nonlinear AL that incorporates the AR components (reprinted with permission from Zagadou et al., 2016).

large enough that the middle ear stiffens and the flow of higherfrequency energy, also present in the sound, is reduced. It also means that the specific time of arrival of oscillations makes a real difference in their effect.

In Zagadou et al. (2016) a new biomechanically based model was developed combining the AHAAH model and integrated cochlear energy (ICE) as the damage risk correlate (DRC) (Fig. 7). The ICE is defined as the sum of the energy accumulated over time in each of the 23 critical bands along the human cochlea, which represents the total energy deposited in the cochlea:

$$ICE = \sum_{i=1}^{23} \int_{T} P_i(t) V_i(t) dt$$
(5)

where T is the duration of the exposure, and  $P_i$  and  $V_i$ , the pressure difference across the BM and the BM volume velocity, for each location, *i*, respectively. The pressure difference across the BM is derived from the impedance and the volume velocity of the BM. The BM impedance is given in terms of the BM resistance, the BM compliance, and the BM mass, which are functions of the position along the cochlea and the angular frequency of the sound pressure. A logarithmic function of the *ICE* was used to express the dosage as dB:

$$Dose = 10 \cdot \log_{10} \left( \frac{ICE}{ICE_0} \right) + 10 \log_{10}(N) \tag{6}$$

where ICE represents the average DRC value at each exposure level. The reference value  $ICE_0$  represents the estimated smallest ICE value possible at  $1.92e^{-15}\mu J$ . For multiple shots, the dose is assumed to accumulate according to the traditional  $10 \log_{10}(N)$ rule, where N represents the number of shots. The model parameters were modified from cat to human based on recent literature data. The unwarned condition was adopted for all the cases presented and the acoustic reflex and annular ligaments nonlinearities were incorporated in the middle-ear model. The new transfer function showed good agreement with the measured transfer functions from cadaver ears and the anomalous dose-response inversion behavior of the original AHAAH model was corrected. The modeling results showed that the annular ligament (AL) parameters were the dominant cause of the non-monotonic doseresponse behavior of AHAAH and a physiological 40% reduction of the AL compliance from the AHAAH default value removes the dose-response inversion problem. Furthermore, the dose-response curve established using the ICE as the DRC and the data from the Walk-up study (Johnson, 1994) showed excellent agreement with rifle noise data (Brinkmann, 2000; Pfander et al., 1980) and with the injury outcomes, hence providing a significant independent validation of the improved model.

## 8. Summary

Significant advances on analytical, physical and numerical models of the hearing system have permitted to better understand the mechanisms behind normal hearing, and in some cases models have also been used for explaining loss of hearing under diverse pathological conditions and exposure to high intensity sound. In this review, we described models of the normal hearing process within the different compartments of the hearing system, including the external, middle, inner ear, and the micromechanics of the organ of Corti, as well as integrated models of the entire hearing chain. Two periods can be clearly distinguished on the modeling of the normal hearing system. From the first half of the 20th century to the end of 1970s, modeling of hearing comprised mostly analytical and lumped models in 1D or 2D. Those early models included a number of simplifications and lacked experimental data for some of the parameters employed, but were in general effective in describing the overall behavior of the process being studied. They lied down the foundation in which more sophisticated models and hypotheses on the function of the hearing system were built. The improvement of computational techniques in the 1980s allowed for the development of finite element and lumped models capable of describing not only the hearing function but to integrate a more refined depiction of the 3D macro/microscopic morphology, material properties, fluid-solid coupling phenomena.

In the 1990's, progress on analytical, physical and numerical models of the external, middle and inner ear have helped increasing our understanding of the normal hearing process, and more recently, several of those models have been extended to the study of the nonlinear behavior of some compartments of the hearing system. Among those improvements we count the use of (1) anisotropy and hyperelastic material properties for the tympanic membrane and other soft tissues, including the behavior of muscles and ligaments on the biomechanics of the middle ear, (2) the nonlinear behavior of the basilar membrane in the cochlea, the outer hair cells and (3) geometrical nonlinearities in the organ of Corti. In few instances, those models have been developed for high intensity noise, for which the nonlinear behavior of tissues and substructures of the ear are critical. Damage and rupture of the tympanic membrane and the ossicular chain has been successfully predicted by some models as well as the pressure distribution inside the ear canal. The inner ear, on the contrary, has received more attention to model the cochlear impedance, the middle ear transfer function, and the basilar membrane displacement and the organ of Corti micromechanics in physiological conditions. Such trend suggests that the modeling field seeks for a more complete understanding of the normal function of the cochlea and the organ of Corti before being able to describe and predict damage and loss of function in the inner ear.

Entire hearing system approaches, in particular the AHAAH model, has undergone a remarkable evolution from the first simplified lumped parameter model proposed 25 years ago and based on cat parameters to our days. The AHAAH model, currently based on human data, is capable of assessing auditory risk units that are converted into compound transient or permanent threshold and thus associated with hearing loss. Auditory hazard has been successfully assessed in the military clinical studies by the AHAAH model for warned and unwarned conditions and free-field and shielded conditions under high-pressure sound energy.

In turn, FE models of the entire hearing system have also been useful for analyzing middle-ear disorders such as otitis media, otosclerosis and ossicular chain disarticulation, and for investigating the effect of tympanic membrane perforation and repair. Numerical models of the whole hearing system have the advantage of providing 3D details of the morphology, which may allow for an accurate patient-specific risk calculation in the future. However, to date, a comprehensive nonlinear FE model of the human hearing system capable of providing an assessment of auditory risk units, threshold shifts and hearing loss due to high intensity sound, is still missing in the literature.

#### **Conflict of interest**

The authors have no conflicts of interest.

#### Acknowledgements

This work was supported by QUASAR (TO 0036), NSF (CMMI-1333560, MRI-0723027, and MRI-1229449), and NIH (DK103362).

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