Prediction of airborne and structure borne sound transmission through hearing protectors using FEM

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ABSTRACT

Individual hearing protection is a short term solution frequently used to protect workers against noise exposure. This work is part of a larger research program on the evaluation of real world attenuation of hearing protection devices (HPD) such as earplugs and earmuffs. Its objective is to develop a Finite Element (FE) model to predict both the airborne and structure borne sound transmission through an ear canal/hearing protector system. The model can help providing a better understanding of the acoustical behavior of this system and hence help designing better HPD ultimately. FEM can account for realistic ear and hearing protector geometries, materials with complicated physical behavioral laws and complex transmission phenomena through the different coupled domains in particular the tissues surrounding the ear canal. Several comparisons between numerical predictions and experimental results are presented. The sound transmission through the system is investigated using power balances and sensitivity analyses are carried out to identify the key parameters which govern the acoustic behavior of the system. The importance of parameters such as the system geometry, the coupling between the different domains and the loadings is discussed together with the characterization of the physical properties of each domain.

Keywords: Sound, Transmission, Hearing, I-INCE Classification of Subjects Number(s): 75.3,36.1,36.2

1. INTRODUCTION

Approximately 500000 workers in the province of Quebec (Canada) are regularly exposed to noise levels that can permanently damage the auditory system. A widespread solution used to protect the worker from noise overexposure consists in using hearing protection devices (HPD), such as earplugs (EPs) or earmuffs (EMs). Wearing HPD is however associated with several issues such as a risk co-factor in work accidents, insufficient in-situ protection of the worker compared to laboratory conditions, difficulty of using standardized measurement techniques REAT (real ear attenuation at threshold) and IL (insertion loss) in workplaces, new measurement techniques such as F-MIRE (field microphone-in-real-ear) require further research efforts to be used with HPD such as earmuffs and HPD wearing time recommended to limit noise exposure is often not respected because of physical or auditory discomfort problems. Examples of such problems include the effect of the static pressure induced by the earmuff headband force or the occlusion effect (OE) which expresses itself through a distortion of the wearer's own voice and the amplification of physiological noises when the HPD is worn.

Some of these issues have been addressed in a 4 year research project funded by IRSST using both experimental and modeling approaches. The ultimate goal of this research was to better evaluate the objective real world attenuation of HPDs and to help designing more efficient HPDs. A large part of the project was dedicated to the development of individual field attenuation measurement methods and to the measurement of the OE (which may have an impact on the occluded exposure sound level). Another major part of this work aimed at using numerical modeling to achieve an optimal design of a

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field attenuation measurement set-up developed in the scope of this research project by (i) providing a better understanding of the physics of the ear canal/hearing protector system (e.g influence of microphone positioning used to assess the field attenuation of circumaural hearing protectors, sensitivity of the acoustic performance of earmuffs to sound directivity, identification of airborne and structure borne sound transmission paths in the system, effect of the HPD design on its acoustical performance etc.) (ii) helping to interpret experimental results (iii) helping to improve HPD acoustic design. This paper focuses on this modeling work and is a follow up of what has been presented in (1).

This research allowed for developing numerical models to predict the sound transmission through earmuffs and EPs for both airborne and structure borne excitation. Methods and experimental set-ups have also been developed to characterize some of the models physical parameters and to evaluate the models. More details will be provided in Section 2.

A comprehensive literature review until 2012 regarding the modeling of the external auditory system coupled to an HPD was provided in (1). It revealed that developing a robust model allowing for predicting the vibroacoustic response of the full problem/HPD/ear canal/head/torso accounting for both airborne and structure borne excitation remained an open question. Sgard et al (1) discussed related modeling issues such as the coupling between the skin and the EP, the importance of the system geometry, of the biomaterial properties of the human external ear and the vibroacoustic behavior of the earmuff cushion. Recently, Viallet et al(2) discussed the effect of the geometry of a molded silicon EP-EC system embedded in a rigid baffle on EP attenuation. They also investigated the validity of the 2D-axisymmetric hypothesis to predict the EP sound attenuation. They identified conditions in which a 2D axisymmetric EP-rigid walled EC geometry can be used rather than a full 3D geometry. In another paper, Viallet et al carried out the same analysis for an EP inserted into an ear canal with surrounding biological tissues (skin alone (3) or (bone, skin and cartilage) (4)). They also investigated the possibility of replacing the effect of the surrounding tissues (bone and cartilage) with a mechanical impedance boundary condition calculated from the complete model in order to reduce the complexity of the system (5). Brummund et al(6) proposed a 3D linear elasto-acoustic FE model of the ear canal and surrounding tissues to simulate the OE induced by an EP. The model provided OE simulations that were consistent with what had been reported in the literature. The authors continued their work by investigating the influence of the material properties of the external ear on simulated OE data(7). Brummund et al(8) also proposed a simplified artificial test fixture of the external ear to measure the objective OE in order to validate the numerical OE models. Ultimately, this test fixture could be used to better assess the sound attenuation and the OE of EPs and help to improve the acoustic design of EPs. Boyer et al(9) carried out an experimental assessment of the sound paths through two different commercially available earmuffs in order to better understand the vibroacoustic behavior of each component of the earmuffs, and to ultimately identify an adequate modeling approach based on the Finite Element Method (FEM). They also compared different approaches to model the vibroacoustic behavior of earmuffs at low frequency including two models based on the FEM where the cushion was either modeled as a spring foundation (SF) or as an equivalent solid (ES), and an existing Lumped Parametric Model (LPM). They pointed out the importance of characterizing the cushion mechanical parameters as a function of the compression rate of the cushion and frequency.

The goal of the present paper is to provide the reader with an overview of the modeling approaches which have been developed in the framework of the research project mentioned previously. Several examples, mostly derived from the aforementioned articles (1,2,4,6–9) are presented to illustrate the potential of the modeling tools. In these papers, numerical models with increasing complexity have been successively developed and validated by comparisons with experimental results carried out at ICAR laboratories in the École de Technologie Supérieure (ÉTS) Montreal and with data from the literature, when available. The limits of the accompanying assumptions have been assessed by comparing the results of the simplified models to the model with the highest complexity. In all these works, the fluid-structure interaction problem of the HPD either excited acoustically or mechanically was solved using the Finite Element (FE) method in the frequency range [50Hz-5000Hz]. The coupling of the HPD/head/torso system with the surrounding medium was accounted for using Automatically Matched Layer (AML) or simple radiation impedance boundary conditions when relevant. Each component of the HPD together with the ear canal and the surrounding biological tissues were discretized with finite elements. This paper is organized as follows. In section 2, the modeling approaches together with the challenges associated to each component of the problem are recalled. Section 3 presents examples of comparisons between simulation results and experimental data. Finally, conclusions are presented in section 4.

2. MODELING APPROACHES

Only a brief summary of the updated modeling challenges together with the used modeling strategies which have already been detailed in (1), are discussed here.

2.1 Problem of interest

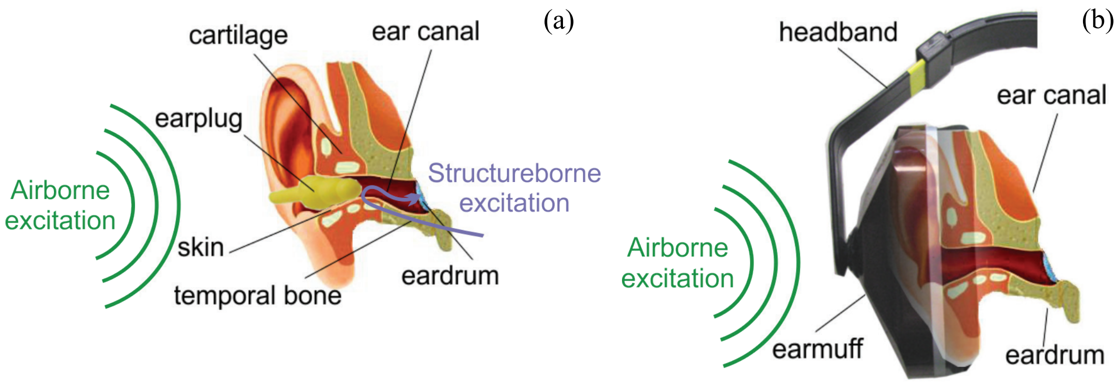

Fig. 1 – Systems of interest. (a) EP coupled to ear canal. (b) earmuff coupled to ear canal.

The systems to be modeled are illustrated in Figure 1. They consist either of an EP inserted in an ear canal (left) or an earmuff coupled to an ear canal (right). The ear canal is surrounded by biological tissues (bone, cartilage, skin) and is terminated by the eardrum. In the frequency range of interest, the eardrum is described by a normal locally reacting acoustic impedance boundary condition obtained from lumped models available in the literature (2,10,11). Earmuffs (see Figure 5) consist of a plastic envelope coupled to an annular polymeric sheath filled with air, comfort foam and in some cases an oil pouch. The interior of the envelope is made up of an air cavity partly filled with an acoustic foam coupled to the cushion air cavity and the ear canal. EPs have various shapes and can be made up of silicone (molded EPs), foam or plastic. These HPDs are assumed to be directly excited by a sound wave (case of EP and earmuff) or indirectly by a mechanical solicitation applied on the bony part surrounding the ear canal (case of EP). This structure borne excitation is used to evaluate the OE. Each component of the HPD together with the ear canal, the surrounding biological tissues and the fluid domain are discretized with finite elements. Either LMS Virtual Lab 12.0 (HPD/EC/head/torso FEM/AML configuration) or COMSOL Multiphysics (HPD/EC configuration) are utilized to solve the FE systems.

2.2 Coupling of the HPD/head/torso with the surrounding medium

The fluid-structure interaction problem of the HPD fully coupled to the head/torso (actually an Acoustical Test Fixture (ATF)) and the acoustic environment is solved using a FE/AML approach in the frequency range [50Hz-5000Hz]. The coupling of the system with the surrounding medium is accounted for using a convex fluid domain which encloses the system and which is connected to the AML. This problem can also be simplified by neglecting the head/torso presence together with the external fluid-structure coupling for both the airborne and structure borne excitation and using instead simple impedance boundary conditions and acoustical loading conditions which apply directly onto the HPD. For example, the ear canal system can be considered as embedded in an infinite rigid baffle. This simplification does not change the acoustic transmission behavior of the HPD and allows for a reduction of the number of degrees of freedom of the system to be solved and faster computation times.

2.3 Ear canal and surrounding biological tissues

The modeling of the ear canal and its surrounding tissues is an important challenge in this research. Full 3D or 2D axisymmetric models have been considered. For structure borne excitation, the surrounding tissues (skin, cartilage, bone) have to be accounted for to model the OE since waves originating from the excitation of the mastoid process propagate in the different tissues and HPD, and end up radiating into the ear canal, thereby inducing an OE upon HPD insertion. There are several difficulties associated with the implementation of such 3D or 2D models: (i) limitation of the region to be modeled (ii) reconstruction of the associated geometry (dimensions and thicknesses of each domain) (iii) boundary conditions and loading to be applied on the domains of interest (ii) selection of behavioral laws of each domain (iv) assessment of the corresponding physical parameters. For airborne excitation, it is also important to account for the biological tissues since the response of the HPD is highly dependent on the HPD/flesh mechanical coupling. Acoustic flanking paths through the flesh can also affect the sound pressure level in the EC depending on the insertion depth of the EP.

In this work, the effect of the skull bones other than the temporal bone and other tissues, such as the brain, was neglected as a first approximation. For the modeling of the OE (structure borne excitation) three models of decreasing geometrical complexity were considered (a) a detailed 3D model of the temporal bone coupled to a cylindrical shape that contains cartilage, skin, EC and tympanic membrane (tissue domain interfaces are of realistic geometry) (b) a second 3D model where tissue domain interfaces are of realistic geometry as well, but where temporal bone is reduced to a more simple cylindrical shape (c) a 2D axi-symmetric model (constant or variable cross section cylinders). For airborne excitation, only models (b) and (c) were retained. The EC and surrounding tissues geometries were either reconstructed from the Visible Human Project® database (NLM, MD, USA) or from data available in the literature (6,12). The different geometrical approximations (3D/2D) and the influence of the surrounding tissues (skin, cartilage, bone) on the sound attenuation were assessed in (2,3,13,14). The sensitivity of the acoustic response to the choice of boundary conditions and loadings was investigated in (13–15) for the OE and in (16) for the sound attenuation. The characterization of the external ear biomaterial properties (bony and cartilaginous tissues, skin) for acoustic applications remains a very challenging task. In this work, biomaterials have been assumed to be isotropic linearly elastic and their properties have been approximated using literature findings (13). Even though this choice of behavioral law and associated properties for biomaterials is debatable, satisfying agreement between simulations with measured data could be obtained. Undeniably, there is a need to elaborate appropriate non-invasive or invasive test protocols to determine the mechanical properties of the bone, skin and cartilage tissues of the external ear. Sensitivity analyses of the calculated response to the geometrical and physical properties of the EC surrounding tissues have been performed to identify the most important parameters (2,3,13,15,16).

2.4 Earplug

The attenuation and the OE of EPs depend mainly on (i) their physical properties, their insertion depth, their mechanical coupling with the EC walls (ii) the leaks caused by an improper insertion of the EP or by the presence of a measurement microphone (F-MIRE) (iii) the excitation. All these factors can be accounted for in the FE model of the fully coupled problem of the EP with the surrounding tissues, the main difficulty being to assess precisely the various input parameters. There remain open issues such as how the EC sound pressure level is influenced by (i) the actual geometrical shapes of the EC and EP when the latter is inserted in the EC since both systems deform depending on their respective stiffness (ii) the associated local changes of EP and EC physical properties due to their interaction. This calls for the development of appropriate test rigs to evaluate these geometrical and physical properties changes and requires to include them adequately in the full model.

In this work, the EP mechanical parameters were either characterized in laboratory using a DMA (silicone EP) or were derived from the literature (17) (foam EP). The coupling between EC walls and EPs (simple kinematic boundary conditions, impedance boundary conditions, full coupling) was studied in (4) for airborne excitation. Sensitivity analyses of the calculated response to the model parameters (geometry, insertion depth, leaks, physical properties of EP, boundary conditions) have been carried out to identify which parameters play an important role and should be known accurately. These analyses also give some insight of which variable should be controlled with care to design efficient EPs. The role of each part of the EP/EC system and how the energy circulates within the domains has also been investigated by calculating power balances together with mechanical and acoustical fluxes in the system (3). It has been assumed that the EC does not deform and that the EP perfectly adapts to the ear canal shape. In addition, the changes of EP global stiffness and damping properties as a function of its insertion depth in the EC were accounted for in the case of the foam EP using data from the literature (17).

2.5 Earmuff

At low frequency, the sound attenuation of EMs is governed by the pumping motion which depends on the headband force, the earcup mass and the stiffness of the cushion/flesh system. In this frequency zone, a lumped model (LPM) or a SF model can be sufficient to predict the acoustic response. However, in (18), it was shown that LPM parameters depending on the EM geometry had to be adjusted to values which do not correspond to those given in the literature, in order to match experimental data. On the contrary, numerical models relying on the true EM geometry, provided excellent agreement with experiments, particularly when the frequency-dependence of the cushion mechanical parameters was accounted for. At high frequencies, the sound transmission is governed by the earcup+EM cavity+foam insert system. In between these two frequency zones, coupling effects occur between the different components of the EM in particular the cushion and the backplate. In addition, sound paths through the cushion and the cup can contribute similarly to the sound attenuation in frequency zones of limited extent (indicatively 250-500Hz) mainly at mid frequencies (typically 2.25-3.5kHz depending on the earmuff type). In order to predict the sound transmission loss through earmuffs in the frequency range [50Hz-5000Hz], it is therefore necessary to model correctly both the mechanical and acoustical behavior of the cushion coupled to the flesh, together with the right couplings between the components (backplate-cushion, earcup-foam insert, cushion-flesh). An associated difficulty lies in the choice of model for the cushion, the characterization of the corresponding physical parameters together with the identification of the coupling conditions between the components.

In this work, complete FE models of two commercial EMs lying on a rigid baffle have been developed based either on a SF or an equivalent isotropic viscoelastic solid cushion model (18) and compared to measured data. A characterization procedure based on a hybrid experimental-numerical inverse approach has also been proposed to identify the cushion frequency dependent and compression rate dependent physical properties. However, the equivalent isotropic viscoelastic solid model for the cushion failed to capture the right sound transmission behavior through the cushion flanks in some frequency zones. An alternative would be to develop a more sophisticated multi-domain model. The influence on the sound attenuation of the coupling conditions between the cushion and the back-plate on the one hand and the foam insert and the earcup on the other hand is currently investigated via a sensitivity analysis (presence of gluing flaws etc.) (19).

3. RESULTS

This section presents some examples of comparisons between experimental data and numerical results obtained with the various developed numerical models. More results will be discussed during the oral presentation.

3.1 Simulation of the sound attenuation through earplugs

The first examples concern the sound attenuation through EPs (which is quantified by the IL) and are taken from (3,16). Figure 2 compares the simulated and measured IL of a 5.8mm long custom made silicon EP inserted in an ATF 45CB without the pinna. The IL is calculated using an axisymmetric FE model depicted in Figure 2. The ear canal and EP are represented as straight cylinders of identical cross-sectional areas. The ear canal walls consist of a cylindrical artificial skin layer which is backed by a steel cylinder in the lateral outer ear (close to ear canal opening) and a steel cylinder in the medial outer ear (close to the eardrum). Note that in the proposed model, the EC is assumed to be embedded in a flat rigid baffle rather than in the ATF. The relevance of this simplification has been examined in (3). Given the small size of the system compared to the acoustic wavelength, a normal incidence plane wave has been used as excitation in the model rather than a diffuse field as in the measurement. The values of the density, Young's modulus, Poisson's ratio and loss factor were 1150 kg.m^{-3} (respectively 1050 kg.m^{-3}), 1.2 Mpa (respectively 0.42 Mpa), 0.48 (respectively 0.43) and 0.12 (respectively 0.2) for the silicon (respectively skin). The numerical results are seen to be in good agreement with the experimental data.

Figure 3 compares the predicted ILs with the ILs measured on human subjects for a 11.7mm long custom molded EP (20). An optimum fit (no leak) and two sizes of leakages are considered. The calculation is based on the average 2D axisymmetric FE model described in (16). Leaks are introduced in the model via a thermally conducting and viscous air cavity through the EP. The predicted mean IL is located in the range of variation of the standard deviations obtained experimentally with just one exception at 1.25 kHz. At low frequency, the introduction of leakages leads to a decrease of the predicted IL. As frequency increases, the IL differences between the model with and without leakages progressively reduce. These leakages can explain the relatively important standard deviation obtained for IL measurements in the low frequency bands ≤1kHz.

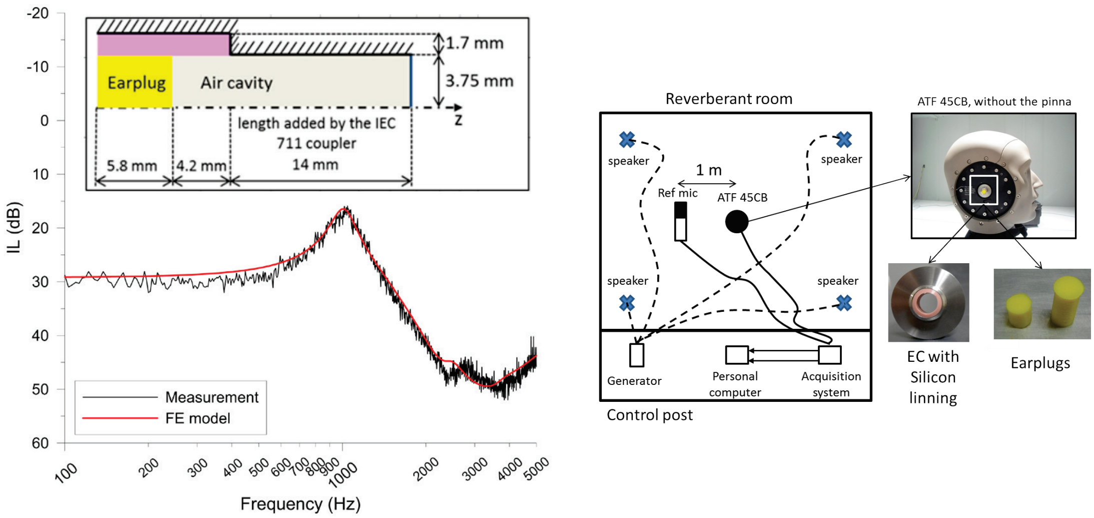

Fig. 2 – Left: comparison of predicted and measured IL for a 5.8mm silicone EP; Right: experimental validation set-up (adapted from (3))

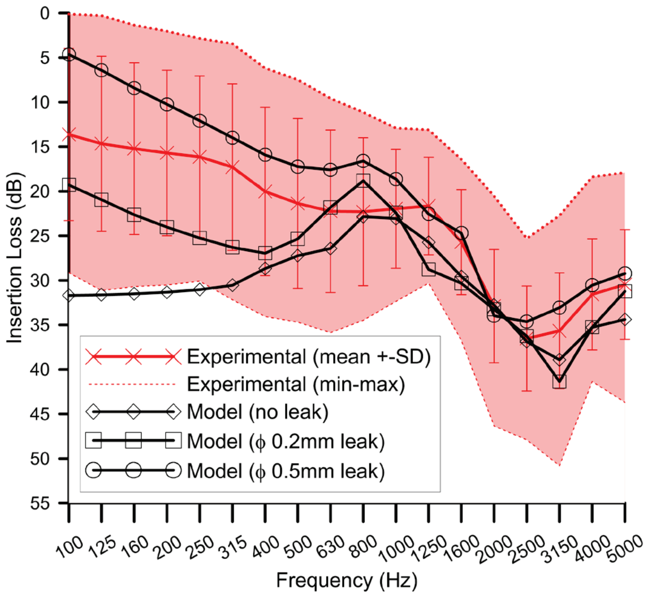

Fig. 3 – Comparison between measured (mean ± SD) and predicted third octave band IL of a 11.7mm long custom molded EP (no leak, 0.2 mm and 0.5 mm diameter leaks) (adapted from (16)).

3.2 Simulation of the occlusion effect

Next example is the simulation of the OE which is derived from (14). The OE is defined as the sound pressure level difference in the EC with and without the EP for a bone conduction excitation.

Figure 4b and Figure 4c display the numerical OE predictions obtained for a foam (11.1mm insertion) and a silicone (11.7mm insertion) EP together with the objective OEs (mean ± S.D.) measured in 37 human subjects at corresponding EP insertion depths (measured with respect to the ear canal entrance) (20). The prediction is based on an average 2D axisymmetric model described in (14) and depicted in Figure 4a. The ear canal and EP are represented as straight cylinders of identical cross-sectional areas. The ear canal walls are formed by a cylindrical skin layer which is surrounded by a cylindrical cartilage layer in the lateral outer ear (close to ear canal opening) and a cylindrical bone layer in the medial outer ear (close to the eardrum). The skin tissue of the lateral ear canal wall in the cartilaginous part is slightly thicker than that in the bony tissue. The values of the density, Young's modulus, Poisson's ratio and loss factor were 1050kg.m^{-3} (respectively 220, 1100, 1080, 1714kg.m^{-3}), 0.85 Mpa (respectively 0.1, 0.5, 7.2, 11316Mpa), 0.48 (respectively 0.1, 0.4, 0.26, 0.3) and 0.1 (respectively 0.5, 0.1, 0.05, 0.01) for the silicon (respectively foam, skin, cartilage, bone). A uniform structure borne mechanical boundary load normal to the horizontal boundaries was applied on the cartilage and bone tissues. The remaining outer tissue boundaries were considered as fixed except for the part of the skin in contact with the external medium, which is free.

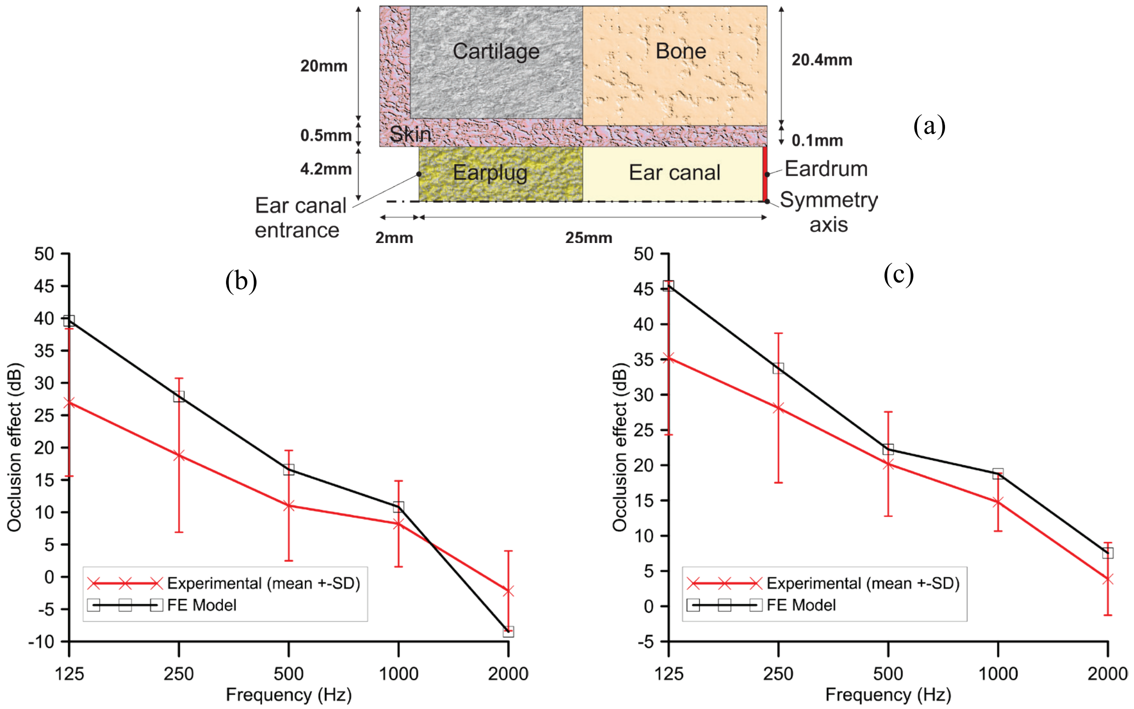

Fig. 4 – Comparison between predicted and measured OE (a) Configuration used in the simulations (b) foam EP (11.1mm insertion depth) (c) silicone EP (11.7mm insertion depth) (adapted from (14))

Figure 4 shows that the predicted OE falls inside the range of the mean ± one standard deviation of the experimental data for both earplug type test groups, albeit close to the upper mean ± S.D boundary in some frequency bands. The low frequency deviation between the numerical and experimental OEs of both EP types could be explained by two reasons. First, the numerical model does not account for EP leaks which can reduce the OE. Second, a low signal to noise ratio due to the airborne sound emission from the bone transducer used in the measurements could have contributed to increase the experimental open ear transfer function level and therefore to decrease the OE.

3.3 Simulation of the sound attenuation through an earmuff

The last example is the simulation of the sound attenuation through a commercially available earmuff (PELTOR-OPTIME 98) (without foam insert) and is taken from (19). The earmuff is assumed to lie on a rigid baffle and to be excited by a blocked pressure field corresponding to an incident plane wave normal to the baffle. The cushion is either modelled as a SF, or as an ES which use frequency dependent equivalent stiffness parameters (linear dependence or 4-parameter fractional derivative of Zener's model) due to the polymeric constitution of its components. In the computations using the cushion ES model, the sound excitation on the cushion's flanks was neglected since this model is not fully adequate to capture the sound transmission through the cushion flanks especially at mid and high frequencies (18). The values of the density, Young's modulus, Poisson's ratio and loss factor were 1040kg.m^{-3} (respectively 1370kg.m^{-3}), 2.2 GPa (respectively 2.4GPa), 0.38 (respectively 0.38) and 0.05 (respectively 0.05) for the earcup (respectively backplate). The cushion density and Poisson's ratio were 70.59kg.m^{-3} and 0.4 respectively. Regarding the linear frequency dependence of the equivalent stiffness $k_{eq}(f) = af + b$ and Young's modulus $E_v(f) = Af + B$ the following coefficients were selected a=336.82, b=32319; A=585.71, B=51393. A constant loss factor equal to 0.22 was used. Regarding Zener's model, the equivalent stiffness (respectively Young's modulus) are obtained from the following expression $\left(M_0 + M_\infty (j\omega\tau_r)^\alpha\right) / \left(1 + (j\omega\tau_r)^\alpha\right)$ where $M_0 = 36633$, $M_\infty = 90338$, $\tau_r = 8.89e^{-4}$, $\alpha = 0.769$ for the equivalent stiffness and $M_0 = 59986$, $M_\infty = 144452$, $\tau_r = 9.51e^{-4}$, $\alpha = 0.805$ for the Young's modulus. The corresponding loss factor is given by

$\eta = \Im(M(j\omega)) / \Re(M(j\omega))$. All coefficients (of both the linear and Zener's models) were obtained from curve fitting based on quasi-static compression measurements carried out for 10 Hz<f<45 Hz.

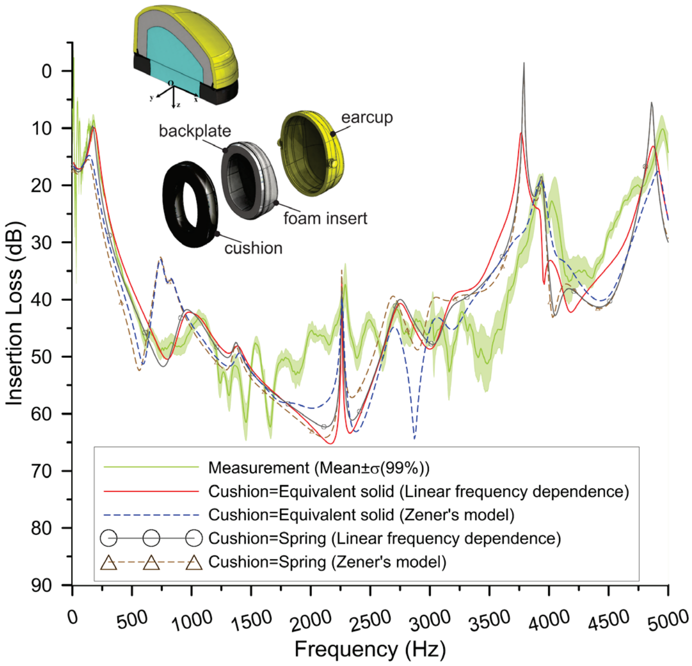

Figure 5 – Comparison between predicted and measured IL for a PELTO-OPTIME 98 earmuff (adapted from (19)).

Figure 5 compares the ILs predicted by the FEM for the aforementioned various cushion models with experimental data. The measurements were carried out in a hemianechoic chamber with a microphone located at the center of the steel baffle " to measure the sound pressure level (SPL) underneath the HPD (9). Overall, all the cushion models allow for capturing the trends of the experimental data except in some isolated frequency bands. Zener's model provides less good results at low frequencies (around 800 Hz) and mid-frequencies (around 2800Hz) but allows to better account for vibration couplings between the cushion and the backplate at higher frequencies (e.g., around 3500Hz and 5200Hz). Both SF and ES model lead to similar results. The various peaks correspond to the pumping motion (at low frequency) and various coupled structural modes of the system governed by the backplate, the earcup or the internal cavity. The differences observed between the numerical simulations and the measurements are attributed to undesired acoustic reflections on the loudspeaker membrane or leakages when carrying out measurements, unsatisfactory cushion model and bad description of the coupling between the backplate and the cushion (19). This work thus highlighted the crucial impact of the cushion vibratory behavior and coupling effects.

4. CONCLUSIONS

The aim of this paper was to provide an overview of numerical modeling strategies which have been developed for predicting the acoustic response of HPDs coupled to the external ear subjected to airborne and structure borne excitations. Examples of comparisons between simulation results and experimental measurements regarding the sound attenuation of EPs and EMs together with the OE of EP, have been presented to illustrate the potential of the various developed modeling tools.

Associated challenges have also been discussed. They include an adequate modeling of the ear canal and its surrounding tissues (including the skull), the selection of the behavioral laws of biomaterials together with the characterization of the corresponding physical parameters, the knowledge of the actual geometrical shapes of the EC and EP when the latter is inserted in the EC and the associated local changes of EP and EC physical properties due to their interaction, the identification of the structure-borne excitation (and possibly the associated airborne noise) to evaluate the OE, the description of the coupling between the EP and the EC walls, a satisfying FE model (multidomains) for EMs cushions accounting for the interaction with the flesh, the knowledge of the coupling conditions between the EM components.

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