AN IN-EAR NOISE DOSIMETRY METHOD THAT EXCLUDES THE SOUNDS GENERATED BY INDIVIDUALS WEARING EARPLUGS: PRELIMINARY FIELD STUDY

Fabien Bonnet
École de technologie supérieure, Montréal, Québec, Canada
email: fabien.bonnet@etsmtl.ca

Hugues Nélisse
IRSST, Montréal, Québec, Canada
email: hugues.nelisse@irsst.qc.ca

Marcos A.C. Nogarolli
École de technologie supérieure, Montréal, Québec, Canada
email: marcos-antonio.coser-nogarolli.1@ens.etsmtl.ca

Jérémie Voix
École de technologie supérieure, Montréal, Québec, Canada
email: jeremie.voix@etsmtl.ca

Every day, hundreds of millions of employees worldwide are exposed to noise levels that are likely to affect their hearing. While noise reduction at the source remains the preferred solution to address the issue of occupational noise, industrial workers are often left with no other option but to wear hearing protection devices (HPD). Unfortunately, the effective protection provided by a given HPD on a given individual depends upon several subject-dependent variables (e.g. earcanal geometry, HPD fitting and wear time). This makes it difficult, with conventional measurement methods, to properly assess the amount of noise an individual receives during a workshift. To overcome this issue, new measurement techniques have been recently developed, such as in-ear noise dosimetry (IEND). IEND may be integrated into HPDs to perform continuous sound pressure measurements under a hearing protector, and hence obtain personal noise exposure assessments accounting for HPD performance. IEND, however, raises questions about the impact of self-generated noise on measurements, as speech or movements (e.g. chewing, walking) from the wearer may significantly affect the sound pressure levels (SPL) measured inside occluded earcanals. As part of this study, an IEND method was developed to perform noise dosimetry measurements under an earplug while excluding the disturbances induced by the wearer. The approach, which was validated in the laboratory using real-ear measurements performed on still human test-subjects, had yet to be tested in real life environments. This paper presents the results obtained on subjects moving in indoor and outdoor urban environments, and provides recommendations about the application of such method in occupational settings.

Keywords: Occupational health and safety, In-ear noise dosimetry, Advanced hearing protection, Occlusion effect, Wearer's own voice

1. Introduction

Personal noise exposure measurements aim to assess the amount of noise exposure for a person, usually a worker, to ensure this amount complies with the exposure limits set by a given legislation. One way of monitoring the level of exposure is a personal body-worn dosimeter, which provides the convenience of continuous monitoring at the location of the individual. Personal noise dosimeters are particularly useful when individuals are required to move frequently during their work shift or when the acoustic environment of the workplace is hardly predictable, since such variables cannot be taken into account with standard sound level meter measurements. These devices are usually attached to the wearer's shoulder to measure the noise levels close to the ears. Though adequate, this location does not always counter the effect of microphone placement, particularly for directional sound fields [1]. Also, the measured sound pressure levels (SPL) may not represent the ambient noise correctly if influenced by the wearer's voice [2], [3]. And furthermore, the accuracy of personal noise dosimeters is compromised when hearing protection devices (HPDs) are worn as the attenuation provided by the HPD (which should be subtracted from the ambient noise levels) can show large variations and uncertainties [4]. Such uncertainties both in the ambient noise levels and in the HPD's effective attenuation make it difficult to accurately determine the actual noise exposure received by a given worker wearing HPDs.

Systems that continuously monitor an individual's noise exposure under the HPD [5]–[8] show promise to remedy these issues. By measuring personal noise exposure directly inside one's occluded earcanals, these may finally provide a clear answer to the pressing question "Is this worker properly protected against noise?" But to do so, the influence of wearer-induced sounds on in-ear noise dosimeter measurements needs to be considered, since the SPLs measured under HPDs may be significantly affected by noise emitted by the wearer. This is particularly true when earplugs are worn, as the so-called occlusion effect (OE) is known to amplify most sounds originating from the wearer, especially at low frequencies. Such sounds, which will be further referred to as wearer-induced disturbances (WIDs), may result from shouting, speaking, singing, coughing or sneezing, but softer sounds associated with chewing, walking, scratching, sniffing, or swallowing may also contribute to the measured SPLs in low ambient noise environments. Besides, research has previously shown that the risk of hearing loss inherent to self-generated noise can be less than that of external noise due to inhibition mechanisms occurring both in the middle ear [9], [10] and at the neuronal level [11]. Moreover, the OE tends to amplify nonphysiological noise emanating from the interaction between the measuring instrument and the wearer, such as rustling and thumping noises (often referred to as microphonics) one hears when tapping the earpiece's cord or when the cord brushes against something. Thus, it is of clear interest to measure the average noise exposure excluding the noise induced by the wearer, especially when earplugs are worn. In a previous paper [12], a low computational method was presented to perform in-ear noise dosimetry under an earplug while excluding WIDs. This method showed good results in laboratory settings involving 14 still participants [12], but should be implemented in the field for further validation. This paper describes preliminary results obtained on moving subjects in indoor and outdoor urban environments to better represent the conditions encountered in typical occupational settings. The method [13] and parameters are presented in the second section. The new results emerging from urban settings are depicted in the third section, together with the experimental setups used. These new results are discussed in the fourth section, which provides recommendations with regard to the application of such method in typical industrial workplaces.

2. Method

2.1 Description

The proposed methodology uses a dual-microphone earpiece, illustrated in Figure 1. An in-ear microphone (IEM) connected to a probe-tube measure the sound pressure under the earplug, while an outer-ear microphone (OEM) measures the sound pressure outside the ear. Although the earpiece can support various types of eartips, the results presented in this paper were obtained using double-flanged silicone eartips, chosen for its easy insertion. A method is proposed to detect and exclude WIDs for dosimetry purposes. The method is based on the following principle: when the sound pressure level measured inside the ear is due to surrounding noise, a strong correlation exists between the two microphone signals as sound simply travels from the OEM to the IEM through the earplug. When the IEM's signal is perturbed by WIDs, such as speech, this correlation drops within the frequency range of the disturbance signal.

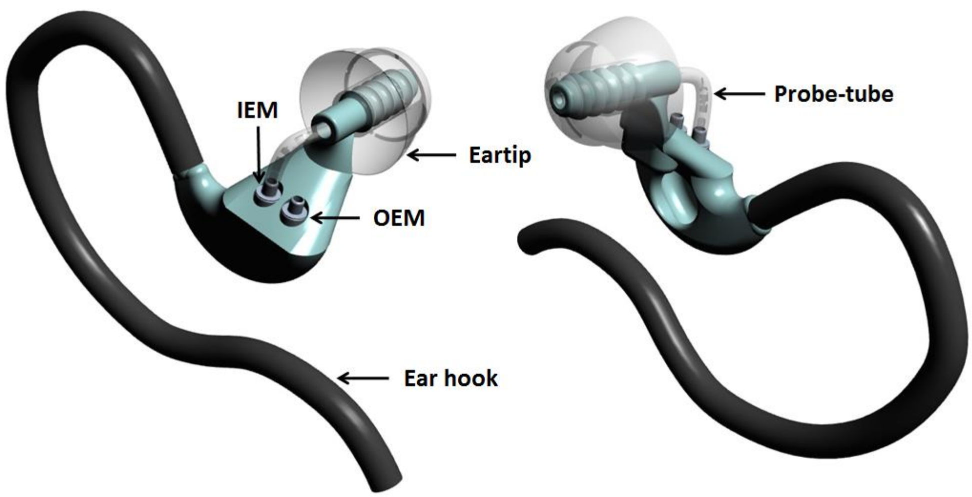

Figure 1: Three-dimensional model of the dual-microphone earpiece used as part of the proposed method

A classical tool to measure the correlation between two signals at specific frequencies is the coherence function γ^2 [14]. It is defined as:

$$\boldsymbol{\gamma^2(f) = \frac{|S_{OI}(f)|^2}{S_{OO}(f)\, S_{II}(f)}} \qquad (1)$$

where $S_{OO}(f)$ is the autospectrum of the time signal *o(t)* measured by the OEM, $S_{II}(f)$ is the autospectrum of the time signal *i(t)* measured by the IEM, and $S_{OI}(f)$ is the cross spectrum between the two signals *o(t)* and *i(t)*. Coherence $\gamma^2(f)$ measures the degree of linear relationship between the two signals at any given frequency or band center frequency, on a scale from 0 (*o(t)* and *i(t)* are uncorrelated) to 1 (*o(t)* and *i(t)* are fully correlated).

For a given time-frame *i*, it is possible to calculate the coherence function at specific frequencies and to average it across the desired frequency range. The indicator Δ is defined here and expressed as follows:

$$\Delta_i = -10 \log_{10}\left(\frac{\sum_{f_p=f_{\min}}^{f_p=f_{\max}} \gamma_i{}^2(f_p)}{N}\right) \qquad (2)$$

where $f_{\min}$ and $f_{\max}$ are the lowest and highest bands of the desired frequency range to be determined, and N is the number of frequency bands within this range. Δ_i, a positive number expressed in dB, approaches 0 when the two microphone signals are highly coherent between $f_{\min}$ and $f_{\max}$, over time frame *i*. The values of $f_{\min}$ and $f_{\max}$ should be representative of the disturbing signals to be detected (e.g., speech). Also, Eqs. (1) and (2) should be implemented as fractional band calculations, since it was found that computing Δ from narrow band values decreased detection performance as it gave too much weight to higher frequencies.

Δ should be computed for every time frame of duration ΔT (e.g., at every 0.5 s), and compared to a threshold value Δ_{th} above which it is assumed a substantial part of the signal measured by the IEM consists of noise contributions from the wearer. When $\Delta_i < \Delta_{\mathrm{th}}$, the impact of WIDs on the sound pressure received by the IEM is negligible, which implies that:

$$\boldsymbol{L^*_{\mathrm{IEM},i}(f) \approx L_{\mathrm{IEM},i}(f)} \qquad (3)$$

where $L_{\mathrm{IEM},i}(f)$ is the SPL measured inside the occluded ear during time frame i, and $L^*_{\mathrm{IEM},i}(f)$ is the SPL that would be measured in the absence of WIDs. When $\Delta_i > \Delta_{\mathrm{th}}$, $L^*_{\mathrm{IEM},i}(f)$ can be estimated using two different methods:

1. $L^*_{\mathrm{IEM},i}(f)$ computed assuming earplug attenuation remains constant during WIDs:

$$\boldsymbol{L^*_{\mathrm{IEM},i}(f) \approx L_{\mathrm{OEM},i}(f) - \mathrm{NR}_{\mathrm{tmp}}(f)} \qquad (4)$$

   where $L_{\mathrm{OEM},i}(f)$ is the SPL measured by the OEM during time frame i, $\mathrm{NR}_{\mathrm{tmp}}(f)$ is the estimated noise reduction (SPL difference between OEM and IEM) measured when the "$\Delta < \Delta_{\mathrm{th}}$" condition was last met, i.e. the last time no WIDs were detected. This approach is particularly adapted for WIDs that increase the SPL inside the ear but barely make any difference to SPL measured outside the ear by the OEM (swallowing, breathing, microphonics, etc.). This method is more adapted to low to medium noise environments as such WIDs, hereafter referred to as "low-level WIDs", typically hardly contribute to the IEM's signal in high noise environments.

2. $L^*_{\mathrm{IEM},i}(f)$ approximated assuming ambient noise levels remain constant during WIDs:

$$\boldsymbol{L^*_{\mathrm{IEM},i}(f) \approx L_{\mathrm{tmp}}(f)} \qquad (5)$$

   where $L_{\mathrm{tmp}}(f)$ is the SPL measured by the IEM when the "$\Delta < \Delta_{\mathrm{th}}$" condition was last met (i.e. the last time no WIDs were detected), or when $L^*_{\mathrm{IEM}}(f)$ was last estimated using Eq. (4). This approach is particularly suited for WIDs that significantly affect the levels measured by the OEM. Such WIDs, hereafter referred to as "high-level WIDs", typically include all vocal WIDs (speech, cough, throat clearing, etc.) as well as the noise resulting from whistling or some shocks to the earpiece. This method is not adapted to low noise environments in which the wearer's physiological noise (breathing, heartbeats, etc.) contribute continuously to the sound pressure inside the earcanal, hence making it difficult to meet the "$\Delta < \Delta_{\mathrm{th}}$" criterion even for short periods of time.

These two methods should be used together as each is adapted to a specific type of WIDs, which implies that a strategy should be found to distinguish low-level WIDs from high-level WIDs. The most obvious characteristic to help differentiating between the two is the in-ear SPL generated by the corresponding signals. Indeed, high-level WIDs such as speech are likely to generate higher in-ear SPLs than low-level (and non-vocal) WIDs. Hence, a simple way to distinguish high-level WIDs from low-level WIDs is to consider a threshold level L_{th} below which no high-level WIDs can theoretically occur. The in-ear SPL in the frequency range of interest is defined as:

$$\boldsymbol{L_i = 10\log_{10}\left(\sum_{f_p=f_{\min}}^{f_p=f_{\max}} 10^{\frac{L_{\mathrm{IEM},i}(f_p)}{10}}\right)} \qquad (6)$$

To be consistent with Eq. (2), the in-ear SPLs and threshold value L_{th} are compared within the same frequency range used to calculate Δ ($f_{\min} < f < f_{\max}$). Whenever $L_i > L_{\mathrm{th}}$, any detected WID is considered as "high-level" (i.e. having a significant impact on $L_{\mathrm{OEM},i}(f)$), which implies that method 2 should be used rather than method 1.

2.2 Parameter values

The method's parameters were optimized using the previous laboratory data [12], and are given in Table 1. These values were found to effectively exclude the WIDs of individuals who remained seated in a reverberant chamber [12]. Such WIDs include: voice sounds (speech, cough, etc.), whistling, chewing, sniffing, swallowing, microphonics, scratching one's face and some shocks to the earpiece.

Table 1: Optimized parameter values selected for the proposed method, where "hangover" refers to a built-in latch mechanism which holds the WID active decision for one sample or more before and/or after detection

f_{min} (Hz)	f_{max} (Hz)	Δ_{th} (dB)	L_{th} (dB)	ΔT (s)	N	*Hangover scheme*
200	1 250	0.75	60	0.3	9	Treat any time frame that precedes or follows a time frame for which the "$\Delta<\Delta_{\mathrm{th}}$" condition is met as if it satisfies the "$\Delta<\Delta_{\mathrm{th}}$" condition itself.

3. Preliminary results in real life settings

The present detection method shows good performance results in laboratory conditions [12]. For a more complete validation, it is essential that the method and algorithms be tested in real-life situations. While a full real-life performance assessment was beyond the scope of this study, the authors were able to collect real-ear measurements in urban environments using the Auditory Research Platform 3 (ARP3). This platform [15], developed by the NSERC-EERS Industrial Research Chair in In-Ear Technologies (CRITIAS), was used together with a 4-channel sound card and is shown in Figure 2. Such measurements allowed testing not only the effects of higher amplitude body movements (e.g. walking), but also the effects of outdoor-specific environmental factors on the measured levels.

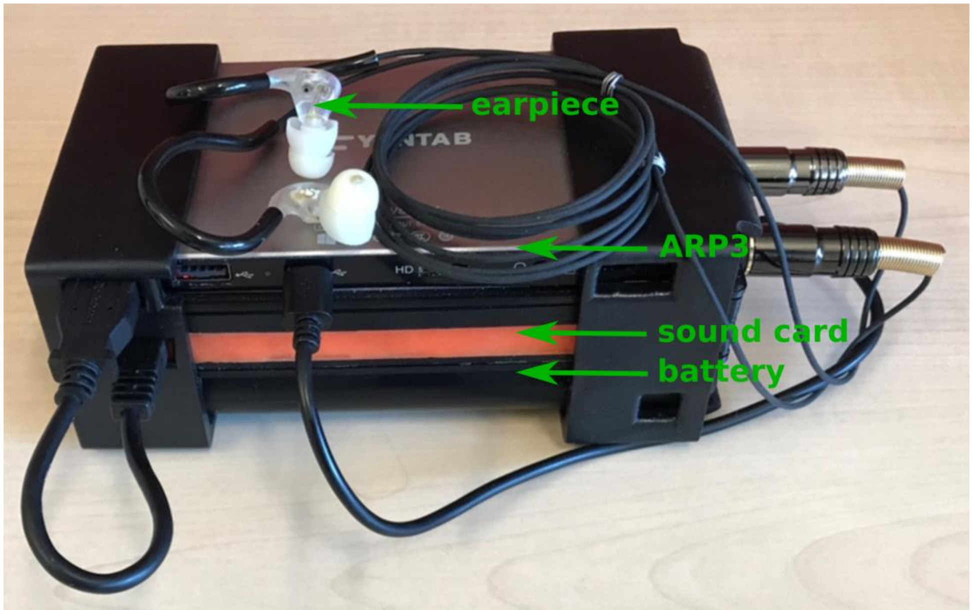

Figure 2: Picture showing the apparatus used for real life testing. The earpieces connect to the ARP3, which is connected to a battery pack providing about 2 hours of autonomy

3.1 Results from indoor measurements

In the laboratory, results were obtained with the subjects seated and the signal conditioning box resting on a support next to them. To provide a baseline for comparison, these conditions were first replicated in a real-life indoor environment. The method was tested in an indoor noisy shopping mall where the wearer remained seated, with the recording platform resting on a table aside. The WIDs tested were the ones mentioned in section 2.2. The results were similar to those obtained in the laboratory.

In industrial workplaces, not only the recording platform shall be worn by the wearer, but the latter may also need to move around with it. This inevitably brings new factors into play, such as noise from footsteps. Indeed, walking with occluded earcanals tends to amplify the wearer's footsteps through bone conduction and induce significant noise levels under the hearing protector, while having a limited impact on the OEM. Data were collected to assess the effect of footsteps' noise on the IEM, and find out whether these disturbances were detected by the proposed method and algorithms. For a worst case test-scenario, the user was wearing boots with thick, rigid soles as these should induce higher bone conduction noise than lighter shoes and are more representative of the safety boots currently worn in the workplace. For a user walking continuously in a quiet environment, the in-ear SPLs were found to range from 45 to 70 dBA over time frames of 0.3 seconds, which corresponds to a range of approximately 48—73 dB for the parameter L. As for the parameter Δ, it was found to stay well above the 0.75 dB threshold and to drop only when the wearer stopped walking. As a result, footsteps are generally detected by the present method and seem to lie somewhere in between high-level and low-level WIDs, if the parameters values of Table 1 are used.

Such measurements also confirmed the impact of microphonics on the results. Movements from the microphone's electrical cords tend to generate noise under the HPD ($55<L<75$), and such movements are usually originated by the wearer. Hence, mobile wearers are more likely to cause microphonics. To limit the impact of microphonics, all measurements presented in this section were made with the cords scotch-taped to the wearer's earlobe, in addition to the ear hooks visible in Figure 1.

3.2 Results from outdoor measurements

A few measurements were also made in outdoor urban environments, with the user standing nearby constructions sites and holding the ARP3 recording platform. The wearer performed all the actions mentioned in section 2.2, in addition to walking. As compared to indoor environments, only one major difference stood out, and it relates to the effects of wind on the results. Indeed, it was observed that strong gusts of wind may have a significant impact both on the OEM and the IEM. And while the first one seemed easy to resolve, the second one may be slightly more problematic.

The effect on the OEM relates to some rumble noise resulting from the direct impact of wind on the microphone's membrane. This is due to the earpiece's design, which exposes the microphone membrane to environmental factors, but a different design may have avoided this by hiding the membrane further inside the earpiece's body. This can be easily done, for example, by using a probed microphone, just like for the IEM, except that the probe tube would connect to the external end of the earpiece. In this study, this problem was solved by applying a thin wind screen on top of the OEM.

The effect on the IEM refers to the bone-conduction noise the microphone may experience when the wind hits the earpiece and the wearer's head. Because such bone-conducted wind noise (BCWN) events do not affect the OEM, they necessarily lead to a drop of coherence between the two microphones, and were found to be often identified as WIDs by the present method. For typical wind speeds (between 15 and 40 km/h), the impact on the IEM generally lied below the threshold of 60 dB, meaning that BCWN fell into the category of low-level WIDs. On more windy days (wind speeds from 50 km/h), however, it was found that BCWN could largely exceed that threshold ($L>70$), hence falling into the category of high-level WIDs.

4. Discussion

One of the downsides of the present method is that, if high-level WIDs happen continuously for long periods of time, it will be assumed that $L^*_{\mathrm{IEM}}(f)$ remains constant and equals $L_{\mathrm{tmp}}(f)$ until one of the two detection parameters (L, Δ) falls below its respective threshold. For instance, if a wearer speaks continuously for 20 seconds and that her/his voice contributes significantly to the SPLs measured by the IEM, the method may assume that the SPL surrounding the wearer during these 20 seconds is the one measured right before vocalization starts. Hence, any ambient SPL variations occurring during that 20s period will be ignored, leading to erroneous $L^*_{\mathrm{IEM}}(f)$ values. In what follows, such errors will be referred to as 'type A errors'. The purpose of the parameter L is to avoid type A errors to happen for lower-level WIDs. That is, whenever WIDs fall below a certain in-ear threshold level (L_{th}), it is assumed that their impact on the OEM should be small enough that the OEM, together with the attenuation, can be used to approximate the in-ear SPLs due to the environment ($L^*_{\mathrm{IEM}}(f)$). Nevertheless, if the threshold L_{th} is too high, there might be WIDs that fall below that threshold but still significantly affect the levels measured by the OEM. This also leads to errors in the values of $L^*_{\mathrm{IEM}}(f)$, further referred to as 'type B errors'. Thus, the threshold value L_{th} controls the balance between type A and type B errors, as a higher L_{th} decreases type A errors but increases type B errors. In the laboratory, measurements were made using constant ambient SPLs, where type A errors cannot exist. Hence, the value of 60 dB for L_{th} was selected only to minimize type B errors. In typical occupational settings, it may be wise to increase the threshold L_{th}, especially for individuals who tend to move a lot in the workplace. Indeed, the laboratory measurements showed that while the impact of microphonics on the IEM could be important ($55<L<75$), they did not seem to affect the OEM down to 50 dB ambient SPLs [12]. Together with footsteps ($48<L<73$), these are two factors that may cause type A errors for mobile wearers. Besides, previous measurements also showed that the WIDs that affect the OEM the most (speech, shocks on earpiece, whistling) usually correspond to L values of at least 75 dB [12]. Hence, considering that all laboratory measurements were made in a highly reverberant room (this tends to increase the impact of WIDs on the OEM), and unless the method applies to low-mobility workers, it may be advisable to increase the L_{th} threshold up to a maximum value of 75 dB. The authors recommend a threshold value of 70 dB, which provided satisfactory results in the urban settings used.

The results from outdoor measurements raise two questions as to the detection of BCWN. Firstly, It is unclear whether BCWN should be included or excluded from noise dosimetry, as these do not originate from the wearer directly but rather from the interaction between the wearer and her/his environment. Secondly, depending on the choice to include or exclude BCWN, two issues may appear: i) if BCWN is to be excluded, the method seems adapted, but the impact of such noise events on the results should be questioned in the case of long and frequent gusts of wind. On particularly windy days, one should keep in mind that the proportion of type A errors may increase significantly as BCWN will fall into the category of high-level WIDs; ii) if BCWN is to be included in the measured noise dose, the method seems less adapted as it does not distinguish between BCWN and WIDs. Whatever the scenario, one way of dealing with BCWN is to reduce it by design, and hence avoid such complicated issues. In this study, a wool hat that covers the ears and head of the wearer was used as a wind screenwas indeed found to cancel the effects of wind on the IEM ($\Delta< \Delta_{\mathrm{th}}$). Such a solution, however, was only tested for moderate wind speeds (approximately up to 50 km/h), and may not be sufficient under more extreme weather conditions. Besides, this is only a temporary solution, and the present approach would surely benefit from a more effective way of dealing with BCWN. A promising avenue, perhaps, may be to develop dedicated algorithms aiming at the detection of BCWN specifically. If BCWN could be distinguished from WIDs, then it would be up to the user or industrial hygienist to include or exclude it from noise dosimetry.

5. Conclusions

A low computational method was presented to perform in-ear noise dosimetry measurements under an earplug, using a dual-microphone earpiece to offer the possibility of excluding the noise disturbances induced by the wearer. This paper compares previous laboratory results to more recent results obtained in real life environments that better represent the conditions encountered in typical workplace settings. In addition to guidelines regarding instrumentation, recommendations were given on the parameter values to be used for better performance in real life settings. The recommended values should better account for the WIDs resulting from movements from the wearer, such as walking. Further work involves the development of a method to detect bone-conducted wind noise so that the method can be effectively used in outdoor environments regardless of weather conditions.

REFERENCES

[1] D. Byrne and E. Reeves, “Analysis of Nonstandard Noise Dosimeter Microphone,” *J Occup Env Hyg*, vol. 5, no. 3, pp. 197–209, 2008.

[2] S. Ryherd, M. Kleiner, K. P. Waye, and E. E. Ryherd, “Influence of a wearer’s voice on noise dosimeter measurements,” *J. Acoust. Soc. Am.*, vol. 131, no. 2, pp. 1183–1193, Feb. 2012.

[3] M. Borgh, F. Lindström, K. P. Waye, and I. Claesson, “The Effect of Own Voice on Noise Dosimeter Measurements: A Field Study in a Day-Care Environment, Including Adults and Children,” in *The 37th International Congress and Exposition on Noise Control Engineering, October 26-29, Shanghai*, 2008.

[4] J. Voix, P. Smith and E. Berger, “Field Fit-Testing and Attenuation Measurement Procedures,” in *The Noise Manual*, 6th ed., Am. Ind. Hyg. Assoc. J., *in press*.

[5] R. Bessette and K. Michael, “Measure and Intervene: An In-Ear Dosimetry Method That Can Change an OSHA Violation - and Internal Attitudes,” *Hear. Rev.*, vol. 19, no. 4, pp. 46–51, 2012.

[6] M. A. Theis, H. L. Gallagher, R. L. McKinley, and V. S. Bjorn, “Hearing protection with integrated in-ear dosimetry: a noise dose study,” in *Proceedings of the Internoise 2012/ASME NCAD meeting August 19-22, 2012, New York*, 2012.

[7] K. Mazur and J. Voix, “A case-study on the continuous use of an in-ear dosimetric device,” *J Acoust Soc Am*, vol. 133, no. 5, pp. 3274–3274, May 2013.

[8] H. L. Gallagher, R. L. McKinley, M. A. Theis, and V. S. Bjorn, “Calibration of an In-Ear Dosimeter for a Single Hearing Protection Device,” AIR FORCE RESEARCH LAB WRIGHT-PATTERSON AFB OH HUMAN EFFECTIVENESS DIRECTORATE, 2014.

[9] E. Borg and S. A. Counter, “The Middle-Ear Muscles,” *Sci. Am.*, vol. 261, no. 2, pp. 74–80, Aug. 1989.

[10] S. Mukerji, A. M. Windsor, and D. J. Lee, “Auditory Brainstem Circuits That Mediate the Middle Ear Muscle Reflex,” *Trends Amplif.*, vol. 14, no. 3, pp. 170–191, Sep. 2010.

[11] O. Creutzfeldt, G. Ojemann, and E. Lettich, “Neuronal activity in the human lateral temporal lobe. II. Responses to the subjects own voice,” *Exp. Brain Res.*, vol. 77, no. 3, pp. 476–489, 1989.

[12] F. Bonnet, H. Nélisse, M. Nogarolli and J. Voix, “In-Ear Noise Dosimetry under Earplug: Method to Exclude Wearer-Induced Disturbances,” *Manuscript submitted for publication.*

[13] École de Technologie Supérieure, "Méthode de mesure de l’exposition sonore effective intra-auriculaire sous un protecteur auditif de type bouchon" U.S. Patent Application No. 62/669.177, 2018.

[14] R. B. Randall, *Frequency analysis*, 3. ed. Naerum: Brüel & Kjaer, 1987.

[15] “Auditory Research Platform (ARP) | CRITIAS.” [Online]. Available: http://critias.etsmtl.ca/the-technology/arp/. [Accessed: 21-Mar-2019].