

Signal processing techniques for continuous monitoring of distortion product otoacoustic emissions in noisy industrial environments

Vincent Nadon, Jeremie Voix

Département de Génie Électrique, École de technologie supérieure, 1100, Notre-Dame Ouest, Montreal, Quebec, H3C 1K3, Canada

Annelies Bockstael, Dick Botteldooren

INTEC, Acoustics Research group, Ghent University, Sint-Pietersnieuwstraat 41, B-9000 Ghent, Belgium

Summary

Current progress in occupational health to prevent occupational noise-induced hearing loss (NIHL) is based on an average exposure/damage relationship. While this average is established on a group of exposed workers, individual susceptibility to NIHL is an important factor in a worker's actual risk of developing NIHL and is not accounted for at the moment. The measurement of otoacoustic emissions (OAE) is often considered a reliable tool to detect the onset of NIHL. Nevertheless, individual field measurement of OAEs on industrial workers is very challenging in practice because high ambient noise levels hinder proper OAE measurements. Use of OAE measurement probes with high passive noise isolation attenuates most of the high frequency ambient noise, but is often insufficient for the low frequency content. In previous research work by the authors, a new type of OAE system was designed to monitor OAE levels continuously on an individual worker. This new OAE system is capable of detecting a slight change in hearing sensitivity. The system features a pair of intra-aural hearing protectors (earplugs) which are equipped with an external microphone, an internal microphone and a pair of miniature receivers. Adaptive filtering on the OAE microphone signal is used to further improve the Signal-to-Noise ratio of the distortion product OAE (DPOAE) in frequencies where passive isolation remains insufficient. In this study, simulation and benchmarking of a further improved version of a DPOAE signal extraction algorithm designed for small digital signal processors (DSP), against other proven algorithms, shows a more stable performance across various DPOAE signal dynamic ranges and noise floor levels.

PACS no. 43.58.+z, 43.60.+d, 43.64.Jb

1. Introduction

Despite the efforts deployed with workplace hearing conservation programs [1], occupational hearing loss remains a problem. The passive noise reduction of the hearing protectors worn during the work shift greatly differs from the attenuation specification printed on the box due to suboptimal placement of protectors on/in the ear, inconsistent use and general variations in the acoustical seal over time [2]. Despite the recent development of a field attenuation measurement system for hearing protection devices [3, 4], the precise residual noise level under the hearing protector remains unknown [5]. Even if these individual noise exposure were known precisely for each industrial worker, the effective risk of hearing damage would still remain uncertain given that a worker's susceptibility to noise-induced

hearing loss [6] varies from one person to another for a given noise exposure.

An alternative approach recently proposed by the authors consists in measuring the auditory health changes induced by daily noise exposure on an individual basis and to warn the worker immediately when a change in hearing sensitivity is taking place, before any permanent damage is caused [7, 8]. In clinical practice, a wide range of audiological tests are available to assess hearing status. However, with respect to occupational noise exposure, these tests are not conducted frequently enough for early detection of changes in hearing sensitivity induced by noise exposure, and cannot be carried out in an environment where acoustical and electrical noise levels are too high. Moreover, the whole procedure to monitor a worker's hearing health daily takes too much time for most standard audiological tests and would interfere with the worker's work routine.

The measurement of Distortion Product otoacoustic emissions (DPOAEs) offers a fast, objective and reliable way to detect early signs of noise-induced changes in hear-

ing sensitivity [9]. When two pure tone stimuli of frequencies f_1 and f_2 are sent through the two miniature receivers of the OAE probe with an f_2/f_1 ratio typically around 1.22, a low-level cubic distortion signal at frequency $f_{dp} = 2f_1 - f_2$ is generated by an active non-linear process of the cochlea. This response travels back from the inner ear to the outer ear canal where it can be recorded by the miniature microphone of the OAE probe. When outer hair cells within a frequency specific region inside the cochlea are damaged due to excessive noise exposure for example, the otoacoustic emission level is lower than when the hair cells are fully functional. Therefore OAEs provide an objective indicator of the hair cells health.

As normal DPOAE levels fall between -5 dB to 20 dB sound pressure levels (SPL) [10], proper recording is very vulnerable to interfering background noise [11]. Therefore, no commercial system currently on the market can continuously monitor DPOAEs in-field in a given individual, because the DPOAE signal is disturbed by the background noise.

Although additional passive noise reduction and other hardware improvements might improve the Signal-to-Noise ratio (SNR), studies have shown [12, 13, 14] that in order to extract the level of the DPOAE signal in a noisy environment, a more robust signal processing scheme is needed.

An adaptive filtering algorithm was previously developed by the authors [7, 8] in order to reduce the ambient and physiological noises from the DPOAE signal using three microphones simultaneously: the first microphone capturing the DPOAE signal inside the ear canal, the second microphone placed inside the contra-lateral ear canal and the third microphone placed at the vicinity of the tested ear DPOAE probe to capture the external noise.

In order to extract the level of the DPOAE signal after noise reduction processing, a robust sinusoid signal extraction algorithm is needed. Ziarani [15] has shown a promising approach to extract DPOAE signals without a Fast Fourier Transform (FFT). This approach is more robust to higher noise levels and, since it is not FFT based, it can be used at any stimuli frequencies (f_1 and f_2 respecting the 1.22 frequency ratio) without keeping an integer multiple of the frequency resolution ($\Delta f = f_s/N$) of the FFT. The extended stimuli frequency range capabilities of such algorithm, gives the opportunity for researchers to characterize the cochlea functionality with a finer frequency resolution than FFT based algorithms since stimuli frequencies can be adjusted to any frequency.

Simulations of the algorithm proposed by Ziarani [15], conducted previously by the authors, have shown that the algorithm is sensitive to the adjustment of various parameters i.e. filter adjustments, adaptation step sizes and normalization gains. Therefore, such an algorithm is not practical in order to assess a worker's cochlea functionality in an automatic and autonomous manner if parameters need to be changed constantly.

The aim of the current study is to describe a recently developed algorithm for the extraction of DPOAE levels.

This algorithm was designed in order to be easily implemented in a low processing power digital signal processor (DSP). Similar to Ziarani's algorithm [15], the proposed processing scheme can use an extended range of stimuli frequencies to precisely characterize the cochlea's health. The developed algorithm uses a temporal modulation with amplitude and phase tracking capabilities in order to extract the DPOAE signal level. Within the same algorithm, a noise estimator can be used in order to evaluate the SNR of the DPOAE signal in the DSP. Therefore, no FFT is needed for the calculation of the noise levels, hence a simplified DSP structure can be used.

In this paper, the methods used to conduct the simulations of the signal extraction algorithms and benchmarking are described in the following section. The results obtained with the proposed algorithm and other benchmarked algorithms using various DPOAE signal dynamic ranges and noise floor levels are then compared based on the estimation error and presented in Section 3. Analysis of the results is presented in Section 4 followed by the conclusions of the study.

2. Methods

The proposed temporal modulation algorithm, shown in Fig. 1 and described in the following section, was simulated using synthesized DPOAE signals generated in MATLAB®. The DPOAE signals frequency range was based on a standard clinical system [16] with stimuli ranging from $f_2=6169$ Hz to $f_2=1000$ Hz. Each of the twenty-two DPOAE signals had a steady magnitude and phase for 1.4 seconds. A wide-band (white) noise signal was added to the twenty-two synthesized DPOAE signals to evaluate the estimation error of the proposed algorithm in various noise floor scenarios from -20 to +8 dB(SPL).

The magnitude of each of the twenty-two ($f_{dp1} \dots f_{dp22}$) DPOAE signals was set within a minimum and maximum value defined as the dynamic range of the DPOAE signals (see definition of dynamic range in Fig. 1). Different dynamic ranges of DPOAE signals were tested from [-5;5] dB(SPL) to [-30;30] dB(SPL) to establish the algorithms reliability. A larger dynamic range would be a case where human subjects have higher or lower DPOAEs than average.

The estimation error was calculated as the difference between the simulated (synthesized) DPOAE magnitude and the estimated DPOAE magnitude of the tested algorithms. The Root Mean Square Error (RMSE) [17] was then calculated as an 'average' of estimation errors across the twenty-two tested DPOAE frequencies to benchmark the proposed algorithm with a slightly modified nonlinear adaptive algorithm [15] and an FFT algorithm.

2.1. Proposed Algorithm

An Amplitude Modulation (AM) algorithm was developed [18] in order to estimate the DPOAE signal without being

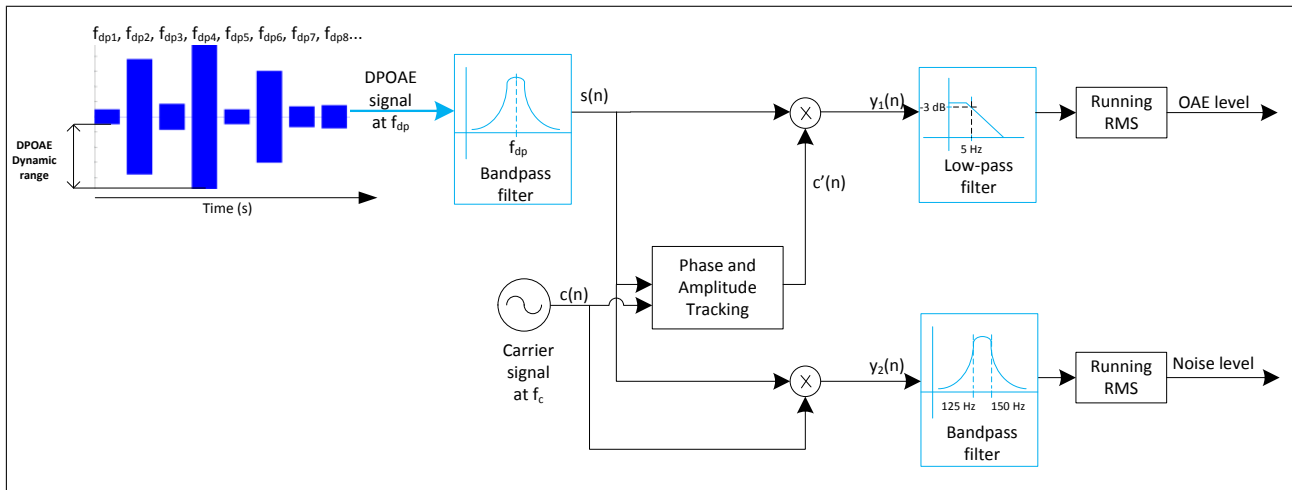


Figure 1. Block diagram of the proposed DPOAE signal extraction algorithm

affected by spectral leakage, which occurs when the stimuli are not an integer multiple of the FFT frequency resolution (Δf). This algorithm has an automatic normalization process that adjusts the modulating carrier signal ($c(n)$) level to match the DPOAE level. Cross-correlation is used to evaluate phase drifts, hence slight frequency variations, to sync the carrier ($c'(n)$) with the DPOAE signal to extract. This way a maximum modulation index (h) is obtained [19], where h corresponds to the ratio between the DPOAE signal magnitude and the carrier signal.

The proposed algorithm consists in the equations shown in the following section (Eq. 1 to 7). The DPOAE signal (see Eq. 1) is modulated with a carrier signal (see Eq. 2) in order to estimate the magnitude of the DPOAE as a constant value (0 Hz). A band-pass Finite Impulse Response (FIR) filter centered around the DPOAE frequency f_{dp} with a filter order $N = 7000$ is used to remove the stimuli signals from the temporal signal prior to the modulation.

Eq. 1 to 7 are executed on a frame of sample size M , this frame size can be set manually or adjusted automatically based on the DPOAE frequency (f_{dp}) in order to reduce the DPOAE stimulation and signal extraction time, especially with higher DPOAE frequencies. In the following study the frame size was adjusted manually to $M = 8000$.

In order to extract the DPOAE signal accurately, the frequency f_c of the modulating carrier $c(n)$ must be the same as f_{dp} the frequency of the DPOAE response signal.

$$s(n) = A_{dp} \sin(2\pi f_{dp} n t_s) \quad (1)$$

$$c(n) = A_c \sin(2\pi f_c n t_s + \phi) \quad (2)$$

To synchronize the carrier signal $c[n]$ with the DPOAE signal $s[n]$, the phase ϕ starts at $\frac{\pi}{2}$ rad and is increased by an additional delay within a loop until the cross-correlation (Eq. 3) gives the closest result to unity. The cross-correlation is used as a measure of similarity be-

tween the DPOAE signal and the carrier signal as a function of the time difference between the signals.

$$(c' \star s)(n) = \sum_{m=0}^{M/2-n} c'(m) s(m+n) \quad (3)$$

A running Root-Mean Square (RMS) value gives the magnitude of the signal over a certain amount of cycles of the sinusoid signal. The RMS value of the signal is calculated with a rectangular window $w(n)$ with length W (Eq. 4).

$$\text{rms}(s(n), w(n)) = \sqrt{\frac{s^2(n) * w(n)}{\sum_{n=0}^W w(n)}} \quad (4)$$

$$c'(n) = \frac{\sum_{n=0}^M \text{rms}(s(n), w(n))}{\sum_{n=0}^M \text{rms}(c(n), w(n))} * c(n) \quad (5)$$

The modulating carrier signal is then normalized ($c'[n]$) based on the RMS value of the DPOAE signal (see Eq. 5). The normalization process maximizes the result of the cross-correlation and the modulation index (h). When this modulation index is maximum ($h = 1$) [19], the modulated DPOAE signal has an optimal dynamic range, which also means that the DPOAE signal estimation error is minimized.

$$y_1(n) = s[n] \cdot c'[n] \quad (6)$$

$$y_1(n) = \frac{A_{dp} A_c'}{2} \sin(2\pi(f_{dp} - f_{c'}) n t_s - \phi) \quad (7)$$

The constant (0 Hz) DPOAE signal obtained in $y_1(n)$ (see Eq. 6 and 7) is then filtered with a low-pass filter to remove the undesired signals such as the residual stimuli signal, noise and the $\sin(2\pi(f_{dp} + f_{c'}) n t_s + \phi)$ component of the modulated DPOAE signal.

The DPOAE level is then estimated by calculating the running RMS value of $y_1(n)$ on one or several test sequences, where a test sequence starts with the first stimuli and ends with the twenty-second stimuli.

2.2. Proposed Noise Estimator

The proposed AM algorithm can also be used as an estimator to evaluate the noise around the DPOAE frequency. The noise estimator output $y_2(n)$ consists in the modulation of the DPOAE signal $s[n]$ with the modulating carrier $c[n]$, $y_2(n) = s[n] \cdot c[n]$. The output $y_2(n)$ then goes through a band-pass filter with cut-off frequencies of 125 Hz and 150 Hz in order to evaluate the noise level between the modulated DPOAE signal and the modulated stimuli signals which are usually at an interval greater than 150 Hz from the DPOAE frequency according to the $f_{dp} = 2f_1 - f_2$. The cut-off frequencies were chosen in a way that the filter order would stay low and that the filter would still remove the DPOAE signal itself (below 125 Hz) from the noise calculations. A running RMS estimates the noise level afterwards. The noise level is calculated as the average plus two standard deviations of the RMS level over time.

2.3. Benchmark algorithms

2.3.1. Nonlinear Adaptive Method

The nonlinear adaptive algorithm proposed by Ziarani [15] was slightly modified for the simulations by removing the two adaptive blocks normally used to extract the stimuli from the DPOAE signal. A band-pass Finite Impulse Response (FIR) filter centered around the DPOAE frequency f_{dp} with a filter order $N = 7000$ was implemented instead in order to minimize the changes in Ziarani's algorithm parameters [15] and therefore reduce the risk of errors. The stimuli were not present in the synthesized DPOAE signals, therefore the band-pass filter was only kept for consistency among the different algorithms and conditions tested.

$$y(n) = A(n)\sin(\phi(n)), \quad (8)$$

$$e(n) = u(n) - y(n), \quad (9)$$

$$A(n+1) = A(n) + \frac{2}{f_s}\mu_1 e(n)\sin(\phi(n)), \quad (10)$$

$$\phi(n+1) = \phi(n) + \frac{\omega_{dp}}{f_s} + \frac{2}{f_s}\mu_2 e(n)A(n)\cos(\phi(n)), \quad (11)$$

The optimal parameters $\mu_1 = 200$ and $\mu_2 = 4000$ used for the DPOAE extraction algorithm proposed by Ziarani (see Eq. 8 to 11 also found in [15]) were established by running several simulations within a loop where μ_1 and μ_2 are changed each time within a range from 20 to 20 000 based on results previously reported in the literature [20].

2.3.2. Fast Fourier Transform Algorithm

An unwrapped FFT algorithm was also used to extract the DPOAE signal level from the synthesized signals. This algorithm targets the DPOAE frequency f_{dp} , calculated from the stimuli frequencies with equation $f_{dp} = 2f_1 - f_2$, by looking at the maximum spectral power within a range of ± 4 Hz. After the DPOAE frequency was identified, the DPOAE level was calculated from the autospectrum and interspectrum values at the corresponding frequency.

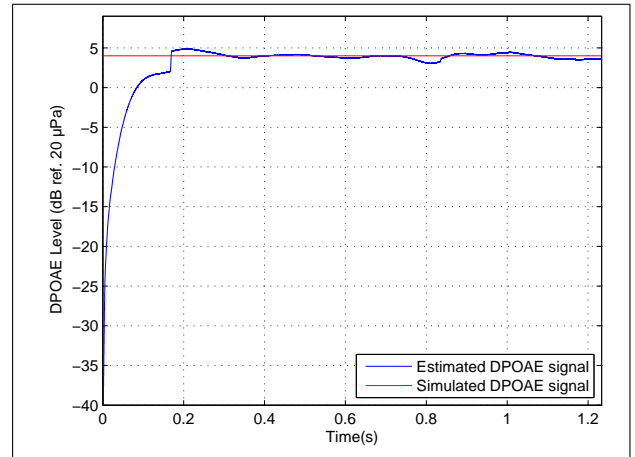


Figure 2. Onset of the estimated DPOAE signal level compared to the original simulated DPOAE level (4 dB(SPL))

3. Results

In order to evaluate the proposed algorithm's performance, a comparative evaluation between the three estimators was conducted using synthesized DPOAE response signals. For a fair comparison of the estimators, only DPOAE responses (not the stimuli) were included in the synthesized signals in order to observe the FFT algorithm results when they are unaffected by spectral leakage (unbiased).

The convergence speed was evaluated to indicate the time required to conduct a full (22 frequencies) DPOAE test. The DPOAE estimation error (RMSE) was observed for the tested estimators in order to establish the algorithms' reliability in various DPOAE signal dynamic range and noise floor conditions.

3.1. Convergence speed

The convergence speed of the proposed algorithm was evaluated by the onset of the estimated DPOAE level as shown in Fig. 2 where the simulated DPOAE signal of 4 dB is compared with the algorithm's output (estimated DPOAE signal). The onset time, which is defined as the time required for the estimated DPOAE signal to reach a stable value within a 1 dB error from the simulated DPOAE signal level, is relatively faster (< 0.2 s) when compared to Ziarani's [15] algorithm (< 0.4 s). Ziarani's method [15] normally uses two additional adaptive processing blocks in order to remove the two stimuli, and therefore needs a time gating of 0.1 s in addition to the 0.4 s convergence time. No time gating process is needed with the proposed algorithm since a band-pass filter removes the stimuli first. As a result, the convergence time obtained with the proposed algorithm is about twice as fast as Ziarani's algorithm [15].

3.2. DPOAE estimation error

The DPOAE level estimation error of each estimator algorithm was compared based on the RMSE value. Results obtained with the proposed algorithm in different noise

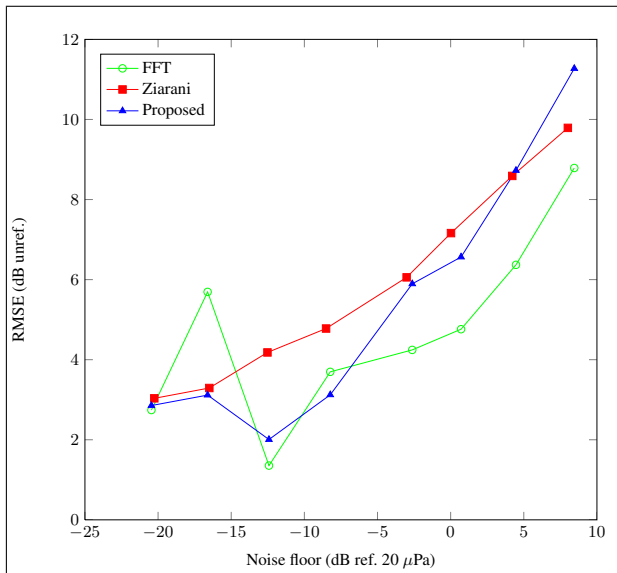


Figure 3. Estimation error evaluated with tested estimator algorithms in different noise levels with DPOAE signals within a range of [-10;10] dB(SPL)

floor conditions ranging from -20 to +8 dB(SPL) with a 4 dB increment (see Fig. 3), where the DPOAE dynamic range is [-10;10] dB(SPL), show that the proposed algorithm is as reliable and robust as Ziarani’s method [15]; except for lower noise floors (around -12 dB(SPL)) where the proposed algorithm has a lower RMSE (2.01 dB) than Ziarani (4.18 dB).

As the DPOAE signal dynamic range is increased, as presented in Fig. 4, the proposed algorithm has a lower and more stable RMSE when compared to Ziarani’s algorithm [15] where the RMSE increases as the dynamic range increases. The RMSE of the FFT estimator is slightly above the RMSE obtained with the proposed method for most of the dynamic ranges tested and it is variable from one test to another. This variability will be discussed in the following section.

4. Discussion

For the purpose of accurate, high frequency and amplitude resolution of DPOAE measurements within a reasonable amount of time, the onset of the DPOAE level and estimation error in different DPOAE signal and noise scenarios are crucial characteristics to evaluate.

The faster convergence speed (0.2 seconds in Fig. 2) of the proposed algorithm will reduce the total measurement time required to less than a half of the usual time required for DPOAE tests with clinical systems [16] on the whole DPOAE frequency range even when some additional samples are recorded for temporal averaging with the proposed algorithm.

As stated in Section 3, the DPOAE level calculated using an FFT based algorithm is highly biased by the spectral leakage in the presence of stimuli signals. This effect was demonstrated by the RMSE average of 5.01 dB

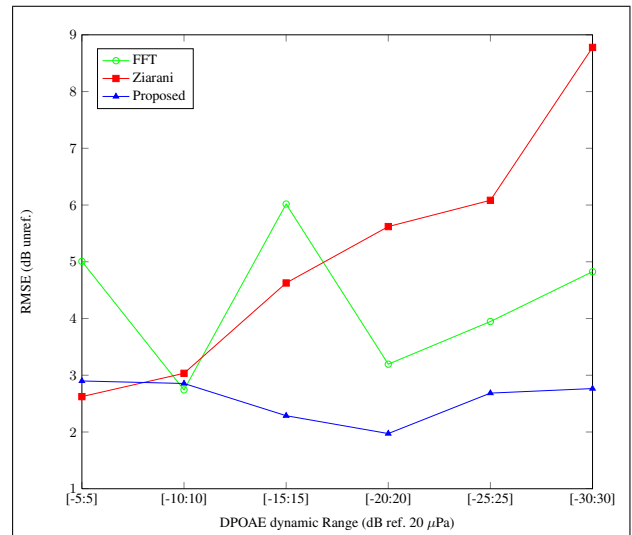


Figure 4. Estimation error evaluated with tested estimator algorithms with different DPOAE signal dynamic ranges and noise floor at -20 dB(SPL)

with 0.44 dB standard deviation observed over ten simulations prior to the study using twenty-two synthesized DPOAE signals within a range of [-10;10] dB(SPL) with stimuli signals ($L_1=65, L_2=55$ dB(SPL)) included and a noise floor of -20 dB(SPL). According to these simulations, the proposed method is less sensitive to spectral leakage since the RMSE stays around 3 dB, which is lower than the RMSE found for the FFT (stable around 5 dB). Hence, the proposed approach is not sensitive to the frequency resolution of the system when extracting the DPOAE signal with the presence of stimuli signals as opposed to FFT based methods commonly used in commercial systems [16]. Therefore, the proposed approach can be used to measure more DPOAE frequencies within a defined range, giving more information about the cochlea’s health throughout the audible frequency range.

Fig. 3 and 4 indicate that the FFT has a somewhat high variability from one test to another, where the performance of other algorithms seems stable and consistent among the tested conditions and also within the same test condition. After running ten simulations of the twenty-two DPOAE frequencies with the FFT algorithm exclusively in the same test conditions ([-10;10] dB(SPL) dynamic range with -20 dB(SPL) noise floor), the average RMSE of 4.99 dB with a 2.02 dB standard deviation confirmed the high variability and that therefore, the FFT is an unreliable estimator. This high variability could explain the need to average the FFT result over several loops observed in previous studies [14].

The proposed algorithm is slightly more robust (lower estimation error) than Ziarani’s algorithm [15] in lower noise conditions (see Fig. 3). Overall, the proposed algorithm has the most stable RMSE in different DPOAE dynamic ranges (see Fig. 4). Ziarani’s algorithm [15] would not be practical in a field application considering the fact that it is very sensitive to the DPOAE dynamic range and therefore various parameters i.e. μ_1, μ_2 , normaliza-

tion gain, number of coefficients and cut-off frequencies of filters need to be adjusted continuously to measure DPOAE's accurately. On the other hand, the proposed algorithm needs no adjustment and still gives a lower RMSE (see Fig. 4). Therefore, it can be fully autonomous, robust and accurate in a field application. The low estimation error of the proposed algorithm will especially be helpful to detect slight DPOAE level variations, hence monitor the cochlea's health status accurately.

5. Conclusions

A newly developed algorithm using temporal modulation with amplitude and phase tracking capabilities for the extraction of DPOAE levels was presented. Inspired by Ziarani's method [15], this algorithm can be used for low processing power DSPs and for measuring accurately an extended range of DPOAE frequencies. Therefore, it is foreseen that in the near future a portable continuous DPOAE monitoring device could be used in the field to closely monitor the cochlea's health in order to warn an industrial worker of a potential change in hearing functionality, hence preventing a permanent hearing threshold shift. Such a device will also help to discover an individual's susceptibility to NIHL as well as the related cochlea health recovery mechanisms to temporary hearing loss.

Acknowledgement

Annelies Bockstael is a postdoctoral fellow of the Research Foundation-Flanders (FWO); the support of this organization is gratefully acknowledged. The ETS-affiliated authors are thankful to the Natural Sciences and Engineering Research Council (NSERC) Individual Discovery Grant Program for part of the funding as well as the technical support from the Sonomax-ETS Industrial Research Chair in In-Ear Technology.

References

- [1] Canetto, Pierre: Hearing protectors: Topicality and research needs. *JOSE* 15.2 (2009), 141-153.
- [2] Néllisse, H., Gaudreau, M.-A., Boutin, J., Voix, J., and Laville, F.: Measurement of Hearing Protection Devices Performance in the Workplace during Full-Shift Working Operations, *Ann. Occup. Hyg.*, **56** (2012) 221-232.
- [3] Voix, J., and Laville, F.: The Objective Measurement of Earplug Field Performance. *J. Acoust. Soc. Am.*, Vol. 125 (2009), 3722-3732.
- [4] Bockstael, A., Botteldooren, D., and Vinck, B.: Verifying the Attenuation of Earplugs in Situ: Comparison of Transfer Functions for HATS and Human Subjects. *Proceedings Inter-Noise* (2007).
- [5] Mazur, K., and Voix, J.: Development of an Individual Dosimetric Hearing Protection Device. *Inter-Noise* (2012). 41th Int. Congr. Expo. Noise Control Eng., 20 p.
- [6] Henderson, D., Subramaniam, M., and Boettcher, F. A.: Individual Susceptibility to Noise-Induced Hearing loss: An Old Topic Revisited. *Ear Hearing*, **14**(3) (1993).
- [7] Nadon, V., Bockstael, A., Keppler, H., Botteldooren, D., Lina, J.-M., and Voix, J.: Use of passive hearing protectors and adaptive noise reduction for field recording of otoacoustic emissions in industrial noise. *Proc. Meet. Acoust., Montreal*, **19** (2013), 040019-040019.
- [8] Nadon, V., Bockstael, A., Botteldooren, D., and Lina, J.-M., Voix, J.: Individual monitoring of hearing status: development and validation of advanced techniques to measure otoacoustic emissions in suboptimal test conditions. *Applied Acoustics* (Submitted 2013). APAC-D-13-00402.
- [9] Marshall, L., Lapsley Miller, J., and Heller, L.: Distortion-product otoacoustic emissions as a screening tool for noise-induced hearing loss. *Noise Health*, **3** (2001), 43-60.
- [10] Moulin, A.: Influence of primary frequencies ratio on distortion product otoacoustic emissions amplitude. II. Interrelations between multicomponent DPOAEs, tone-burst-evoked OAEs, and spontaneous OAEs. *J. Acoust. Soc. Am.*, **107**(3) (2000), 1471-1486.
- [11] Popelka, G. R., Karzon, R. K., Clary, R. A.: Identification of noise sources that influence distortion product otoacoustic emission measurements in human neonates. *Ear and hearing*, **19**(4) (1998), 319-328.
- [12] Bockstael, A., Botteldooren, D., Keppler, H., Degraeve, L., and Vinck, B.: Hearing protectors and the possibility to detect noise-induced hearing damage using otoacoustic emissions in situ. *InterNoise* (2012).
- [13] Bockstael, A., Keppler, H., and Botteldooren, D.: Improved hearing conservation in industry: More efficient implementation of distortion product otoacoustic emissions for accurate hearing status monitoring. *Proc. Meet. Acoust., Montreal*, **19** (2013), 040018-040018.
- [14] Delgado, R. E., Ozdamar, O., Rahman, S., and Lopez, C. N.: Adaptive noise cancellation in a multimicrophone system for distortion product otoacoustic emission acquisition. *IEEE Trans. Biomed. Eng.*, **47** (2000), 1154-64.
- [15] Ziarani, A. K., and Konrad, A.: A novel method of estimation of DPOAE signals. *Conf. Proc. IEEE Eng. Med. Biol. Soc.*, **1** (2004), 380-3.
- [16] Otodynamics Ltd (n.d.). DPEchoport, Available: http://www.otodynamics.com/Product_Echoport.asp?sub=0.
- [17] Willmott, Cort J., and Kenji Matsuura: Advantages of the mean absolute error (MAE) over the root mean square error (RMSE) in assessing average model performance. *Climate Research* 30.1 (2005), 79-82.
- [18] Nadon, V., Voix, J., Bockstael, A., Botteldooren, D., and Lina, J.-M.: METHOD AND DEVICE FOR CONTINUOUS IN-EAR HEARING HEALTH MONITORING ON A HUMAN BEING. Provisional US Patent Application filed through Equinox IP Protection agents, Montreal, Canada, June (2014).
- [19] Klein, H. J., Engineers, C., Francisco, S., Rau, M. C., and Liaison, S.: Modulation, Overmodulation, and Occupied Bandwidth. *National Association of Broadcasters* (1986).
- [20] Wu, Y.: LabVIEW-based GUI for DPOAE Signal Estimation. *Clarkson University* (2006), Master Thesis, p. 27.

FORUM ACUSTICUM
KRAKÓW, 7-12 SEPTEMBER 2014

CD: ISSN 2221-3767

