METHOD AND DEVICE FOR CONTINUOUS IN-EAR HEARING HEALTH MONITORING ON A HUMAN BEING

FIELD OF THE INVENTION

[0001] The present invention relates to hearing protection, and more specifically to a method and device for continuous in-ear hearing health monitoring on a human being based on measurements of otoacoustic emissions (OAE).

BACKGROUND OF THE INVENTION

[0002] Occupational hearing loss remains a problem, despite the efforts made by implementing hearing conservation programs in the workplace [1]. The first issue is that the actual passive noise reduction of the hearing protector worn during the work shift greatly differs from the optimal passive noise reduction measured in the laboratory due to suboptimal placement, inconsistent use and in 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]. The second issue is that, even if this individual noise exposure would be known precisely for each worker, the effective risk of hearing damage would still remain uncertain given the difference between worker's susceptibility to develop noise-induced hearing loss [6]

[0003] To address simultaneously these two issues, an alternative approach would consist in measuring the auditory health changes induced by daily noise exposure on an individual basis and to immediately warn the worker (in real-time) when a change in hearing sensitivity is taking place, before any permanent damage is caused. 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 also not sufficiently robust to be carried out in an environment where acoustical and electrical noise intensity 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.

[0004] Indeed, distortion product otoacoustic emissions (DPOAEs) offer an objective, fast and reliable way to detect early signs of noise-induced changes in hearing sensitivity [7]. When two pure tone stimuli, f_1 and f_2 with the f_2/f_1 ratio typically around 1.22, are sent through the two miniature receivers of the otoacoustic emission (OAE) probe, low-level cubic distortion signals (i.e. $f_{dp} = 2f_1\text{-}f_2$) are generated by an active non-linear process inside the inner ear. These signals travel back from the inner ear to the outer ear canal where they can be recorded. If the outer hair cells inside the cochlea of the inner ear are damaged - for instance due to previous excessive noise exposure - the amplitude of DPOAEs is found to be lower than if they would be healthy. As normal DPOAE levels fall between -5 dB and +20 dB sound pressure levels (SPL) [8], proper recording is very vulnerable to interfering background noise [9].

[0005] Nevertheless, various clinical test setups for DPOAEs have been commercially available for more than 15 years, now ranging from standalone all-in-one hand-held devices to more advanced systems with two probe measurement interfaces connected to a personal computer. No commercial system currently on the market can continuously monitor DPOAEs in a given individual, in field conditions, because the normal DPOAE signal, generally at levels between -5 dB to 20 dB sound pressure level (SPL) [10], is disturbed by the background noise and proper recording is very vulnerable to interfering background noise [11].

[0006] Furthermore, 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 noise reduction algorithm was previously developed by the authors [15, 16] 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 outer ear canal, the second microphone placed inside the contralateral ear canal, and the third microphone placed at the vicinity of the tested ear DPOAE probe to capture the external noise.

[0007] In order to extract the level of the DPOAE signal after the noise reduction processing, a robust sinusoid (or tonal) signal extraction algorithm is needed. Previous studies by [17] have shown a promising approach to extract DPOAE signals without the need of 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 frequency (respecting the 1.22 ratio) without having to keep an integer multiple of the frequency resolution (Δf) of the FFT. The extended stimuli frequency range capabilities of such an algorithm, may give the opportunity for researchers to characterize the cochlea with a finer frequency resolution than FFT based algorithms.

[0008] Simulations of the algorithm proposed by [17], conducted previously by the inventors of the present application, have shown that the algorithm is very sensitive to the adjustment of various parameters i.e. filter adjustments, adaptation step sizes and normalization gains. Therefore, such an algorithm may not be practical in order to assess worker's cochlea functionality in an automatic and autonomous manner if parameters need to be changed constantly.

[0009] In the case of hardware solutions, standard probe eartips usually provide a certain amount of passive noise reduction, but this noise reduction is not individually optimized and it is not sufficient for noisy test environments. Even though passive earmuffs could be used on top of the DPOAE probe, as in [18], they may not provide sufficient additional low frequency attenuation in order to measure DPOAEs accurately in industrial environments. Unfortunately, placing an earmuff on top of an OAE probe might slightly dislocate the probe and hence require more strict supervision of calibration procedures. This situation would conflict with the final aim of OAE monitoring without any external supervision.

[0010] In the case of software solutions, the standard noise rejection techniques and time averaging can improve the signal-to-noise ratio (SNR) in case of limited disturbance [19], but this has shown to be insufficient in more realistic occupational noise settings [20]. Moreover, these techniques do not offer sufficient improvement to lower the noise floor in the frequency range below 1500 Hz to measure DPOAEs accurately even in lower background noise levels [19].

[0011] In response to the problems encountered with averaging methods, several adaptive filtering techniques have been studied [19, 21, 22, 23, 24]. Delgado's [19] adaptive filtering technique uses a contralateral internal ear microphone (IEM) as a physiological noise reference and an ipsilateral outer ear microphone (OEM) as an external background noise reference to remove the noise captured in the tested ear IEM. This adaptive filtering algorithm was proven to increase the SNR on the whole frequency spectrum while reducing the test time needed by normal time averaging methods. Although Delgado's [19] adaptive filtering does lower the noise level in the DPOAE signal, it has not been tested in realistic noise conditions and a somewhat low signal-to-noise ratio improvement was obtained with laboratory setup experiments.

[0012] While the noise reduction algorithm and the involved hardware does lower the noise floor and increase the DPOAE level reliability, it is still necessary to find an alternative to FFT based DPOAE level extraction because the FFT is very sensitive to background noise in frequency bins near the DPOAE frequency and the stimuli. The magnitude of the stimuli and DPOAE response causes spectral leakage around the DPOAE frequency which introduces an error in the estimation of the DPOAE level when the DPOAE frequency is not an integer multiple of the frequency resolution (Δf) [25]. To assess this problem Ziarani [26] proposed a method to extract nonstationary sinusoids with a non FFT based algorithm, but the method is sensitive to the adjustment of various parameters.

[0013] According to the results obtained in previous studies [19, 26], there is a need for an improved method and device for continuous in-ear hearing health monitoring on a human being based on measurements of otoacoustic emissions (OAE) in order to use such device for in-field applications.

SUMMARY OF THE INVENTION

[0014] It is therefore a general object of the present invention to provide an improved method and device for continuous in-ear hearing health monitoring on a human being based on measurements of otoacoustic emissions (OAE), that substantially solve the above-mentioned problems and drawbacks.

[0015] An advantage of the present invention is that the method and device allow a precise real-time variation assessment in hearing status of a given worker through the development of an in-ear hearing protection device - in-ear meaning positioned in the ear canal - featuring in-field otoacoustic emission (OAE) monitoring, more specifically the measurement of Distortion Product OAE (DPOAEs).

[0016] Another advantage of the present invention is that the method and device use an adaptive filtering algorithm consisting in a cascaded two stage adaptive algorithm with a fixed parallel filter. This approach aims to improve DPOAE detection by using three microphones simultaneously: the tested ear internal microphone, the contralateral internal microphone and the reference microphone mounted flush on the outer faceplate of the earpiece-embedded OAE probe. For the adaptive filtering algorithm, two new techniques are used in the present invention: first the influence of the external microphone position is accounted for in order to improve signal denoising; secondly, an online fixed filter is implemented to characterize the primary path transfer function in the adaptive filtering algorithm and the use of a normalized version of the Least-mean square (LMS) algorithm to again improve signal denoising.

[0017] Yet another advantage of the present invention is that the method and device extract nonstationary sinusoids with a temporal modulation algorithm that extracts DPOAE signal magnitude, rather than with a FFT based algorithm. The combination of the adaptive filtering algorithm with the DPOAE signal extraction technique of the present application is new compared to actual DPOAE processing methods.

[0018] Yet another advantage of the present invention is that the method and device use a DPOAE measurement system capable of (a) achieving accurate DPOAE response estimation in (b) elevated background noise by combining improved signal detection algorithms using an advanced noise reduction approach, at least with background noise fragments between about 65 and about 75 dB(A).

[0019] A further advantage of the present invention is that the method and device use an algorithm for the extraction of DPOAE levels that can be easily implemented in a low cost digital signal processor (DSP) with simplified structure, and using a temporal modulation with amplitude and phase tracking capabilities in order to accurately 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. The processing scheme, without FFT, can use an extended range of stimuli frequencies to precisely characterize the cochlea's health.

[0020] Still another advantage of the present invention is that the method and device use a pair of earpieces, typically in-ear devices at least partially located into the outer ear canals, and preferably custom-fitted in-ear devices occluding the outer ear canals as described in U.S. Patent No. 7,864,972 to McIntosh et al. issued on January 4th, 2011, combining simultaneous hearing protection and otoacoustic emissions. Custom-fitted in-ear devices are preferably considered since they provide maximum passive attenuation of ambient noise and reduced placement and fit variability, therefore well suited for enabling continuous monitoring of one's hearing health status, in ‘real-world’ noisy industrial environment, and prevent noise-induced hearing loss.

[0021] According to an aspect of the present invention there is provided a device for continuous in-ear hearing health monitoring on a human being, the device comprising:

- a pair of earpieces, each said earpiece having an internal microphone (IEM-I, IEM-C) adapted to be in fluid communication with an outer ear canal of an ear of a user, an outer ear microphone (OEM-I, OEM-C) adapted to be in fluid communication with an adjacent environment outside the ear of the user, and first and second receivers adapted to be in fluid communication with the outer ear canal;

- a controller system connecting to both the first and second receivers of at least one said pair of earpieces for simultaneously sending first ($f_1(n)$) and second ($f_2(n)$) stimuli sound signals thereto, respectively, the controller system connecting to the internal (IEM-I, IEM-C) of both said earpieces and to the outer ear (OEM-I) microphone of at least one said pair of earpieces for recording respective inner ($d_1(n)$, $x_1(n)$) and outer ($x_2(n)$) sound signals therefrom; and
- an output media connected to the controller system for providing a resulting magnitude of a distortion product otoacoustic emission (DPOAE) signal calculated by the controller system using a signal denoising algorithm and a signal extraction algorithm based on the first ($f_1(n)$) and second ($f_2(n)$) stimuli sound signals and a carrier signal ($c(n)$) of the DPOAE.

[0022] According to another aspect of the present invention there is provided a method for continuous in-ear hearing health monitoring on a human being based on measurements of distortion product otoacoustic emissions (DPOAE) and using a pair of earpieces, each said earpiece having an internal microphone (IEM-I, IEM-C) adapted to be in fluid communication with an outer ear canal of an ear of a user, an outer ear microphone (OEM-I, OEM-C) adapted to be in fluid communication with an adjacent environment outside the ear of the user, and first and second receivers adapted to be in fluid communication with the outer ear canal, the method comprising the steps of:

- simultaneously sending first ($f_1(n)$) and second ($f_2(n)$) stimuli sound signals to the first and second receivers of one of the pair of earpieces, respectively;
- simultaneously recording respective inner sound signals ($d_1(n)$, $x_1(n)$) from the internal (IEM-I, IEM-C) of said earpieces and outer sound signal ($x_2(n)$) from outer ear (OEM-I) microphones of said one of the pair of earpieces; and

- processing the inner ($d_1(n)$, $x_1(n)$) and outer ($x_2(n)$) sound signals of said pair of earpieces to obtain a resulting distortion product otoacoustic emission (DPOAE) based on the first ($f_1(n)$) and second ($f_2(n)$) stimuli sound signals and using a signal denoising algorithm and a signal extraction algorithm.

[0023] According to another aspect of the present invention there is provided a signal denoising algorithm for use in a device or method for continuous in-ear hearing health monitoring on a human being based on measurements of distortion product otoacoustic emissions (DPOAE) and using first and second custom-fitted earpieces being worn by a user, each said earpiece having an internal microphone (IEM-I, IEM-C) adapted to be in fluid communication with an outer ear canal of an ear of a user, an outer ear microphone (OEM-I, OEM-C) adapted to be in fluid communication with an adjacent environment outside the ear of the user, and first and second receivers adapted to be in fluid communication with the outer ear canal, first ($f_1(n)$) and ($f_2(n)$) second stimuli sound signals being simultaneously respectively sent to the first and second receivers of the first earpiece, first inner ($d_1(n)$) and outer ($x_2(n)$, $x'_2(n)$) sound signals being respectively simultaneously recorded from the internal (IEM-I) and outer ear (OEM-I) microphones of the first earpiece, and second inner sound signal ($x_1(n)$) being simultaneously recorded from the internal microphone (IEM-C) of the second earpiece, the signal denoising algorithm comprising:

- a first normalized least-mean square (NLMS) adaptive filtering of the first inner sound signal ($d_1(n)$) to get a first-stage noise reduced DPOAE signal ($e_1(n)$) using a first output ($y_1(n)$) being the filtered second inner sound signal ($x_1(n)$); and
- a second normalized least-mean square (NLMS) adaptive filtering of the first-stage filtered signal ($e_1(n)$) using a second output ($y_2(n)$) being the filtered first outer sound signal ($x_2(n)$) to get a second-stage noise reduced signal ($e_2(n)$) of a distortion product otoacoustic emission (DPOAE).

[0024] In one embodiment, the second normalized least-mean square (NLMS) adaptive filtering includes adaptive filtering of the first-stage filtered signal

($e_1(n)$) using a fixed filtering of the first outer sound signal ($x'_2(n)$) and a second output ($y_2(n)$) of the first outer sound signal ($x_2(n)$) to get a second-stage noise reduced signal ($e_2(n)$) of a distortion product otoacoustic emission (DPOAE), the fixed filtering accounting for a characterization of a primary path transfer function of the first earpiece between the first internal (IEM-I) and outer ear (OEM-I) microphones.

[0025] According to another aspect of the present invention there is provided a signal extraction algorithm for use in a device or method for continuous in-ear hearing health monitoring on a human being, the device or method providing for a filtered signal ($s(n)$) of a distortion product otoacoustic emission (DPOAE), the signal extraction algorithm comprising:

- a temporal modulation of the filtered signal ($s(n)$) of the distortion product otoacoustic emission (DPOAE) with a normalized modulating carrier signal ($c'(n)$) having amplitude and phase tracking capabilities, based on a carrier signal ($c(n)$) being used to set a modulated filtered DPOAE signal as a constant magnitude signal ($y_1(n)$).

[0026] In one embodiment, the signal extraction algorithm further includes a temporal modulation of the filtered signal ($s(n)$) of the distortion product otoacoustic emission (DPOAE) with the modulating carrier signal ($c(n)$) to get a magnitude of a noise signal ($y_2(n)$) around the DPOAE at a predetermined frequency of stimuli sound signals.

[0027] Other objects and advantages of the present invention will become apparent from a careful reading of the detailed description provided herein, with appropriate reference to the accompanying drawings.

BRIEF DESCRIPTION OF THE DRAWINGS

[0028] Further aspects and advantages of the present invention will become better understood with reference to the description in association with the following Figures, in which similar references used in different Figures denote similar components, wherein:

[0029] **Figure 1** is a schematic block diagram of a device and method for continuous in-ear hearing health monitoring on a human being in accordance with an embodiment of the present invention, further showing a signal denoising algorithm in accordance with an embodiment of the present invention and used in the device or method; and

[0030] **Figure 2** is a schematic diagram of a DPOAE signal extraction algorithm use in a device or method for continuous in-ear hearing health monitoring on a human being in accordance with an embodiment of the present invention.

DETAILED DESCRIPTION OF THE INVENTION

[0031] With reference to the annexed drawings the preferred embodiment of the present invention will be herein described for indicative purpose and by no means as of limitation.

[0032] With reference to **Figure 1**, there is shown a schematic block diagram of a device and method for continuous in-ear hearing health monitoring on a human being in accordance with an embodiment of the present invention, and having a signal denoising, or adaptive filtering, algorithm in accordance with an embodiment of the present invention and used in the device or method. The device includes an earpiece typically occluding the outer ear canal, such as a custom-fitted earpiece body as an example, with an earpiece-embedded OAE probe. Each earpiece, a pair of which is schematically shown in **Figure 1**, includes two preferably high-quality miniature balanced armature receivers used to send the two pure-tone stimuli (f_1 and f_2 with the f_2/f_1 ratio typically around 1.22) without any sound distortion from the Auditory Research Platform (ARP). One miniature internal ear microphone, IEM-I, IEM-C, is placed towards the ear canal in order to measure the otoacoustic emission response and physiological noise, respectively, and a miniature outer ear microphone, OEM-I, OEM-C, is placed on the outside of the earpiece to measure the external background noise (see **Figure 1**). The two earpieces are typically connected to a signal conditioning circuit (not shown), part of a controller system, to amplify and filter the DPOAE microphone signal for the digital signal processor (DSP – also part of the controller system) based data acquisition and digital signal processing circuit.

[0033] In the earpiece, the tubing guiding the acoustical signal from the receivers/ microphone to/from the eartip fitting ring ensures that acoustical crosstalk is eliminated as much as possible and background noise is reduced. Self-inflating custom molded earpieces fitted to ensure one's respective ear canals are preferably sealed or occluding the outer ear canal (maximum passive attenuation from the earpiece) for proper DPOAE measurements and also to protect the human subject when measuring in higher levels of background noise.

[0034] When tests with human subjects are being carried out on one ear of the subject, the corresponding earpiece is used to measure the DPOAE (low-level cubic distortion signals (i.e. $f_{dp} = 2f_1\text{-}f_2$) generated by an active non-linear process inside the inner ear) in that ear with the ipsilateral in-ear microphone (IEM-I) and capture the external noise with the ipsilateral outer ear microphone (OEM-I). Meanwhile, the other earpiece is used to capture the physiological noise inside the other ear canal. It is assumed that the physiological noise is similar for both ear canals and since the other contralateral IEM-C does not capture the DPOAE responses evoked at the tested ear but only noise in the other ear canal, therefore IEM-C serves as a noise reference.

[0035] The adaptive filtering or signal denoising algorithm presented in **Figure 1** ensures that the noise measured in the DPOAE response when measuring DPOAEs in noisy conditions is reduced. A bandpass filter is typically used to filter out the stimuli signal from the tested ear's IEM-I to help the adaptive filters converge on the DPOAE signal. Note that the same filter is also typically applied on all other microphones in order to keep the same noise disturbance reference.

[0036] The normalized version of the Least-mean square (NLMS) algorithm adaptive filter in Stage 1 models the transfer function between the tested ear IEM-I $d_1(n)$ and the contralateral IEM-C $x_1(n)$. The output of this filter $y_1(n)$ is then subtracted from the desired signal input $d_1(n)$. The error signal $e_1(n)$ is used to correct the adaptive filter's coefficients in order to model the physiological noise disturbance in the tested ear's IEM-I accurately. In the 2nd adaptive noise reduction stage (Stage 2), $\hat{H}(z)$ is calculated off-line. It models the earplug transfer function ($H(z)$). The adaptive filter is used to compensate for slight variations in the fixed $\hat{H}(z)$ estimated transfer function over time due to variations in earplug seal. Although the fixed filter $\hat{H}(z)$ is shown in parallel in **Figure 1**, it is typically used to initialize the adaptive filter's coefficients in Stage 2 at system's start and can be kept or removed during further testing of DPOAE. Therefore it can be used and illustrated as part of the adaptive filtering block itself when it is used for initialization only. The input of the fixed filter $x'_2(n)$ and the adaptive filter $x_2(n)$ is the band-pass filtered OEM signal. Each output $y'_2(n)$ and $y_2(n)$ is then subtracted from the tested ear's IEM-I to remove the noise disturbance.

[0037] The step size used to adjust the coefficients in the adaptive filters to process the data is set manually per subject within a range of $0.01 \leq \mu \leq 0.5$ for the first stage of the proposed adaptive filtering algorithm (see **Figure 1**) and the second stage was kept at $\mu = 0.01$. Only when more effect of the adaptive filtering algorithm is typically necessary for extreme cases, the step size is then increased up to $\mu = 0.5$. In general, a small step size makes the adaptive filter converge slowly, but the transfer function will be more precise, thus less error is introduced in the DPOAE signal. When the ambient noise has a large dynamic range or high sound pressure level, a higher step size is necessary in order to make the adaptive filter converge in a reasonable time to account for fast variations in the transfer function, but this higher step size may lead to larger DPOAE estimation errors. The step size range is typically limited, especially for the second stage of the adaptive filtering algorithm, in order to not overcompensate the DPOAE response.

[0038] The above 2-stage adaptive filtering algorithm is efficient to denoise the OAE signals for industrial background noises up to about 110 dB(A), and typically at least about 80 dB(A).

[0039] Now referring more specifically to **Figure 2**, there is shown a schematic diagram of a DPOAE signal extraction algorithm use in a device or method for continuous in-ear hearing health monitoring on a human being in accordance with an embodiment of the present invention. In **Figure 2**, the $s(n)$ signal represents the signal coming out of the adaptive filtering algorithm, shown in dashed lines, of **Figure 1**.

[0040] The temporal modulation, more specifically an amplitude modulation (AM) algorithm estimates the DPOAE signal without being affected by spectral leakage which occurs when the stimuli are not an integer multiple of the frequency resolution (Δf) of the FFT. This algorithm has an automatic normalization process that adjusts the modulating carrier signal level to match the DPOAE level. A cross-correlation is used to evaluate phase drifts, hence slight frequency variations, to sync the carrier with the DPOAE signal to extract. This way a maximum modulation index (h) is obtained, thus maximizing the DPOAE level dynamic range and minimizing the Root Mean Square Error (RMSE) of the estimated signal. The RMSE is the average of the estimation errors between the “true” level and the estimated level calculated across the twenty-two tested DPOAE frequencies used in order to benchmark the proposed algorithm with other estimator algorithms.

[0041] The present signal extraction algorithm, in being less sensitive to the frequency resolution of the system to extract the DPOAE signal with the presence of stimuli signals, can be used to measure more DPOAE frequencies within a defined range, therefore giving more information about the cochlea's health throughout the audible frequency range. This increase in DPOAE frequency resolution is especially useful for close monitoring of inner ear health changes in order to understand the recovery mechanisms of the human ear to a temporary hearing loss.

[0042] The present signal extraction algorithm consists in the equations shown in the following section (Equations 1 to 7). The DPOAE signal (see Equation 1) is modulated with a carrier signal (see Equation 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. The Equations 1 to 7 are executed on a frame of sample size (M), this frame size can be fixed 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.

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

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

[0043] To synchronize the carrier signal $c[n]$ with the DPOAE signal $s[n]$, the phase φ starts at $\pi/2$ and is increased by an additional delay within a loop until the cross-correlation (Equation 3) gives the closest result to unity. The cross-correlation is used as a measure of similarity between the DPOAE signal and the carrier signal as a function of the time difference between the signals.

$$(c' * s)(n) = SUM\_\{m=0\}\hat{}\{m=M/2-n\}(c'^*(m)s(m + n)) \quad (3)$$

[0044] 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 (Equation 4).

$$\mathrm{rms}(s(n),w(n)) = \mathrm{SQRT}\ [(s^2(n)w(n)) / \mathrm{SUM}\_\{n=0\}\hat{}\{W\}(w(n))] \quad (4)$$

$$c'(n) = [[\mathrm{SUM}\_\{n=0\}\hat{}\{M\}\ \mathrm{rms}(s(n),w(n))] / [\mathrm{SUM}\_\{n=0\}\hat{}\{M\}\ \mathrm{rms}(c(n),w(n))]]*c(n) \quad (5)$$

[0045] The modulating carrier signal is then normalized $c'[n]$ based on the RMS value of the DPOAE signal (see Equation 5). The normalization process maximizes the result of the cross-correlation and the modulation index (h) which corresponds to the ratio between the DPOAE signal magnitude and the carrier signal. When the modulation index is maximum ($h = 1$), the modulated DPOAE signal has an optimal output, which means that the DPOAE signal estimation error is minimized.

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

$$y_1(n) = (A_{dp}A_{c'}/2)\sin(2\pi(f_{dp} - f_{c'})nt_s - \varphi) \quad (7)$$

[0046] The constant (0 Hz) DPOAE signal obtained in $y_1(n)$ (see Equations 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 $\sin(2\pi(f_{dp}+f_{c'})nt_s+\varphi)$ component of the modulated DPOAE signal.

[0047] 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.

[0048] The 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 about 125 Hz and about 150 Hz in order to evaluate the noise level between the modulated DPOAE signal and the modulated stimuli signal which are usually at an interval greater than 150Hz from the DPOAE frequency according to the $f_{dp} = 2f_1\text{-}f_2$. The cut-off frequencies are 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.

[0049] According to simulations, the present DPOAE signal extraction algorithm is reliable and robust (lower estimation error), mostly for lower noise floors (-12 to -8 dB(SPL)) down to a RMSE of about 2.0 dB. The present DPOAE signal extraction algorithm also has a low and stable RMSE when the dynamic range of the DPOAE signal to extract is increased. The RMSE of the present method is quite consistent over the tested ranges ([-5;5] to [-30;30] dB(SPL)) and lower than any known FFT RMSE for most of the dynamic ranges tested.

[0050] According to previous simulations, the present signal extraction method is more immune to spectral leakage (RMSE around 3 dB for the present algorithm vs RMSE around 5 dB for the FFT). Therefore, it can be used as an estimator when stimuli frequencies and DPOAE frequencies are not an integer multiple of the systems frequency resolution (Δf) and cause spectral leakage.

Thus, the present algorithm can be used to detect changes in the cochlea's functionality with a greater frequency resolution (extended range of DPOAE's) than FFT based methods commonly used in commercial systems since the stimuli frequencies can be set without restrictions.

[0051] The present signal extraction algorithm is also slightly more robust than known algorithms in lower noise conditions. Overall, the present algorithm has the most stable RMSE in different DPOAE dynamic range. In addition, the present algorithm needs no adjustment and still gives a lower RMSE than known algorithms, therefore it can be fully autonomous, robust and accurate in a field application.

[0052] Finally, a faster convergence time (about 0.2 s) of the present signal extraction algorithm reduces the total measurement time required to about a half, for the whole DPOAE frequency range, when compared to known algorithms.

[0053] A present algorithm using temporal modulation for the extraction of DPOAE levels is designed for low cost DSPs and can be used to measure an extended range of DPOAE frequencies. This way a portable DPOAE measurement system can be used to characterize the functionality of the inner ear more precisely. The lower RMSE of the present algorithm is especially helpful to detect slight changes in the cochlea's health accurately. Therefore, a portable continuous DPOAE monitoring device including the present algorithm can be used in field to detect slight changes in the cochlea's health in order to warn the wearer of a potential change in hearing functionality, hence preventing a permanent hearing threshold shift.

[0054] Although the present invention has been described with a certain degree of particularity, it is to be understood that the disclosure has been made by way of example only and that the present invention is not limited to the features of the embodiments described and illustrated herein, but includes all variations and modifications within the scope of the invention as hereinafter claimed.

LIST OF REFERENCES

[1] Canetto, P., 2009. Hearing protectors: Topicality and research needs. JOSE 15, 141-153.

[2] Nélisse, H., Gaudreau, M., Boutin, J., Voix, J., Laville, F., 2011. Measurement of hearing protection devices performance in the workplace during full-shift working operations. Ann. Occup. Hyg., 221-232.

[3] Voix, J., Laville, F., 2009. The Objective Measurement of Earplug Field Performance. Journal of the Acoustical Society of America Vol. 125, 3722-3732.

[4] Bockstael, A., Botteldooren, D., Vinck, B., 2007. Verifying the Attenuation of Earplugs in Situ; Comparison of Transfer Functions for HATS and Human Subjects, in: Proceedings of Inter-Noise 2007.

[5] Mazur, K., Voix, J., 2012. Development of an Individual Dosimetric Hearing Protection Device, in: Inter-Noise 2012 : The 41th International Congress and Exposition on Noise Control Engineering, p. 20 p.

[6] Henderson, D., Subramaniam, M., Boettcher, F., 1993. Individual susceptibility to noise-induced hearing loss: an old topic revisited. Ear Hearing 14, 152-168.

[7] Marshall, L., JA, L., Heller, L., 2001. Distortion-product otoacoustic emissions as a screening tool for noise-induced hearing loss. Noise and Health 3, 43-60.

[8] Moulin, A., 2000. Influence of primary frequencies ratio on distortion product otoacoustic emissions amplitude. ii. interrelations between multicomponent dpoaes, tone-burst-evoked oaes, and spontaneous oaes. The Journal of the Acoustical Society of America 107, 1471.

[9] Popelka, G., Karzon, R., Clary, R., 1998. Identification of noise sources that influence distortion product otoacoustic emission measurements in human neonates. Ear and hearing 19, 319.

[10] Moulin, A. (2000). Influence of primary frequencies ratio on distortion product otoacoustic emissions amplitude. II. Interrelations between multicomponent DPOAEs, tone-burst-evoked OAEs, and spontaneous OAEs. The Journal of the Acoustical Society of America, 107(3), 1471-1486.

[11] Popelka, G. R., Karzon, R. K., Clary, R. A. (1998). Identification of noise sources that influence distortion product otoacoustic emission measurements in human neonates. Ear and hearing, 19(4), 319-328.

[12] Bockstael, A., Botteldooren, D., Keppler, H., Degraeve, L., and Vinck, B. (2012). “Hearing protectors and the possibility to detect noise-induced hearing damage using otoacoustic emissions in situ ,” InterNoise.

[13] Bockstael, A., Keppler, H., and Botteldooren, D. (2013). “Improved hearing conservation in industry: More efficient implementation of distortion product otoacoustic emissions for accurate hearing status monitoring,” 19, 040018–040018.

[14] Delgado, R. E., Ozdamar, O., Rahman, S., and Lopez, C. N. (2000). “Adaptive noise cancellation in a multimicrophone system for distortion product otoacoustic emission acquisition,” IEEE Trans. Biomed. Eng., 47, 1154–64.

[15] Nadon, V., Bockstael, A., Keppler, H., Botteldooren, D., Lina, J.-M., and Voix, J. (2013). “Use of passive hearing protectors and adaptive noise reduction for field recording of otoacoustic emissions in industrial noise,” Proc. Meet. Acoust., Montreal, 040019–040019.

[16] Nadon, V., Bockstael, A., Botteldooren, D., and Lina, 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.

[17] Ziarani, a K., and Konrad, A.: A novel method of estimation of DPOAE signals. Conf. Proc. IEEE Eng. Med. Biol. Soc., 1 (2004) 3803.

[18] Bockstael, A., Botteldooren, D., Keppler, H., Degraeve, L., Vinck, B., 2012. Hearing protectors and the possibility to detect noise-induced hearing damage using otoacoustic emissions in situ. InterNoise.

[19] Delgado, R., Ozdamar, O., Rahman, S., Lopez, C., 2000. Adaptive noise cancellation in a multimicrophone system for distortion product otoacoustic emission acquisition. Biomedical Engineering, IEEE Transactions on 47, 1154-1164.

[20] Bockstael, A., Keppler, H., Botteldooren, D., 2013. Improved hearing conservation in industry: more efficient implementation of Distortion Product Otoacoustic Emissions for accurate hearing status monitoring, in: Proceedings ICA, Montreal.

[21] Ma, W., Zhang, Y., 2002. Estimation of distortion product otoacoustic emissions. Biomedical Engineering, IEEE Transactions on 46, 1261-1264.

[22] Ma, W.K., Zhang, Y.T., Yang, F.S., 1996. Adaptive Filtering for Distortion Product Otoacoustic Emissions. 18th Annual International Conference of the IEEE Engineering in Medicine and Biology Society, 1524-1525.

[23] Ozdamar, O., Delgado, R.E., Rahman, S., Lopez, C., 1998. Adaptive Wiener filtering for improved acquisition of distortion product otoacoustic emissions. Annals of biomedical engineering 26, 883-91.

[24] Ozdamar, O., Delgado, R., 1998. Otoacoustic emission acquisition using adaptive noise cancellation techniques. Proceedings of the 1998 2nd International Conference Biomedical Engineering Days, 139-143.

[25] Ma, W.K., Zhang, Y.T., 1999. Estimation of distortion product otoacoustic emissions. IEEE transactions on bio-medical engineering 46, 1261-4.

[26] Ziarani, A.K., Konrad, A., 2004. A novel method of estimation of DPOAE signals. Conference proceedings : Annual International Conference of the IEEE Engineering in Medicine and Biology Society. Conference 1, 380-3.

CLAIMS

We claim:

1. A device for continuous in-ear hearing health monitoring on a human being, the device comprising:

- a pair of earpieces, each said earpiece having an internal microphone (IEM-I, IEM-C) adapted to be in fluid communication with an outer ear canal of an ear of a user, an outer ear microphone (OEM-I, OEM-C) adapted to be in fluid communication with an adjacent environment outside the ear of the user, and first and second receivers adapted to be in fluid communication with the outer ear canal;
- a controller system connecting to both the first and second receivers of at least one said pair of earpieces for simultaneously sending first ($f_1(n)$) and second ($f_2(n)$) stimuli sound signals thereto, respectively, the controller system connecting to the internal (IEM-I, IEM-C) of both said earpieces and to the outer ear (OEM-I) microphone of at least one said pair of earpieces for recording respective inner ($d_1(n)$, $x_1(n)$) and outer ($x_2(n)$) sound signals therefrom; and
- an output media connected to the controller system for providing a resulting magnitude of a distortion product otoacoustic emission (DPOAE) signal calculated by the controller system using a signal denoising algorithm and a signal extraction algorithm based on the first ($f_1(n)$) and second ($f_2(n)$) stimuli sound signals and a carrier signal ($c(n)$) of the DPOAE.

2. A method for continuous in-ear hearing health monitoring on a human being based on measurements of distortion product otoacoustic emissions (DPOAE) and using a pair of earpieces, each said earpiece having an internal microphone (IEM-I, IEM-C) adapted to be in fluid communication with an outer ear canal of an ear of a user, an outer ear microphone (OEM-I, OEM-C) adapted to be in fluid communication with an adjacent environment outside the

ear of the user, and first and second receivers adapted to be in fluid communication with the outer ear canal, the method comprising the steps of:

- simultaneously sending first ($f_1(n)$) and second ($f_2(n)$) stimuli sound signals to the first and second receivers of one of the pair of earpieces, respectively;
- simultaneously recording respective inner sound signals ($d_1(n)$, $x_1(n)$) from the internal (IEM-I, IEM-C) of said earpieces and outer sound signal ($x_2(n)$) from outer ear (OEM-I) microphones of said one of the pair of earpieces; and
- processing the inner ($d_1(n)$, $x_1(n)$) and outer ($x_2(n)$) sound signals of said pair of earpieces to obtain a resulting distortion product otoacoustic emission (DPOAE) based on the first ($f_1(n)$) and second ($f_2(n)$) stimuli sound signals and using a signal denoising algorithm and a signal extraction algorithm.

3. A signal denoising algorithm for use in a device or method for continuous in-ear hearing health monitoring on a human being based on measurements of distortion product otoacoustic emissions (DPOAE) and using first and second custom-fitted earpieces being worn by a user, each said earpiece having an internal microphone (IEM-I, IEM-C) adapted to be in fluid communication with an outer ear canal of an ear of a user, an outer ear microphone (OEM-I, OEM-C) adapted to be in fluid communication with an adjacent environment outside the ear of the user, and first and second receivers adapted to be in fluid communication with the outer ear canal, first ($f_1(n)$) and ($f_2(n)$) second stimuli sound signals being simultaneously respectively sent to the first and second receivers of the first earpiece, first inner ($d_1(n)$) and outer ($x_2(n)$, $x'_2(n)$) sound signals being respectively simultaneously recorded from the internal (IEM-I) and outer ear (OEM-I) microphones of the first earpiece, and second inner sound signal ($x_1(n)$) being simultaneously recorded from the internal microphone (IEM-C) of the second earpiece, the signal denoising algorithm comprising:

- a first normalized least-mean square (NLMS) adaptive filtering of the first inner sound signal ($d_1(n)$) to get a first-stage noise reduced DPOAE signal ($e_1(n)$) using a first output ($y_1(n)$) being the filtered second inner sound signal ($x_1(n)$); and

- a second normalized least-mean square (NLMS) adaptive filtering of the first-stage filtered signal ($e_1(n)$) using a second output ($y_2(n)$) being the filtered first outer sound signal ($x_2(n)$) to get a second-stage noise reduced signal ($e_2(n)$) of a distortion product otoacoustic emission (DPOAE).

4. The signal denoising algorithm of claim 3, wherein the second normalized least-mean square (NLMS) adaptive filtering includes adaptive filtering of the first-stage filtered signal ($e_1(n)$) using a fixed filtering of the first outer sound signal ($x'_2(n)$) and a second output ($y_2(n)$) of the first outer sound signal ($x_2(n)$) to get a second-stage noise reduced signal ($e_2(n)$) of a distortion product otoacoustic emission (DPOAE), the fixed filtering accounting for a characterization of a primary path transfer function of the first earpiece between the first internal (IEM-I) and outer ear (OEM-I) microphones.

5. A signal extraction algorithm for use in a device or method for continuous in-ear hearing health monitoring on a human being, the device or method providing for a filtered signal ($s(n)$) of a distortion product otoacoustic emission (DPOAE), the signal extraction algorithm comprising:

- a temporal modulation of the filtered signal ($s(n)$) of the distortion product otoacoustic emission (DPOAE) with a normalized modulating carrier signal ($c'(n)$) having amplitude and phase tracking capabilities, based on a carrier signal ($c(n)$) being used to set a modulated filtered DPOAE signal as a constant magnitude signal ($y_1(n)$).

6. The signal extraction algorithm of claim 5, further including a temporal modulation of the filtered signal ($s(n)$) of the distortion product otoacoustic emission (DPOAE) with the modulating carrier signal ($c(n)$) to get a magnitude of a noise signal ($y_2(n)$) around the DPOAE at a predetermined frequency of stimuli sound signals.