

Individual monitoring of hearing status: development and validation of advanced techniques to measure otoacoustic emissions in suboptimal test conditions

Vincent Nadon^a, Annelies Bockstael^b, Dick Botteldooren^b, Jean-Marc Lina^a, Jérémie Voix^{a,*}

^aÉcole de technologie supérieure, 1100, Notre-Dame Ouest, Montreal, Quebec, Canada, H3C 1K3

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

Abstract

Scientific evidence suggests that measurement of otoacoustic emissions (OAE) is a reliable tool to detect the early onset of noise induced hearing loss. Nevertheless, the individual field measurement of otoacoustic emissions on industrial workers is very challenging in practice because the high level of ambient noise usually disturbs the OAE measurement. The use of OAE measurement probes with high passive noise isolation allows the attenuation of most of the high frequency ambient noise, but it is often insufficient for the low frequency content of the ambient noise. In the described research, a new type of OAE system, suitable for the continuous monitoring of OAE levels on an individual worker, has been designed as a pair of earpieces each featuring an external microphone, an internal microphone and a pair of miniature receivers. Adaptive noise reduction (ANR) processing the measured distortion product OAE (DPOAE) is used to further improve the Signal-to-Noise ratio in frequencies mostly where passive isolation remains insufficient. The ANR technique relies on a Normalized Least-Mean-Square (NLMS) algorithm that uses the ipsilateral external microphone and the contralateral internal microphone to denoise the measured DPOAE signals for each in-ear OAE probe. A side-by-side comparison with commercially available clinical OAE equipment on 8 test-subjects successfully confirmed that the developed OAE system would be suitable for the continuous monitoring of workers' hearing capabilities in industrial noise environments with levels up to 75 dB(A).

Keywords: noise-induced hearing loss, otoacoustic emissions, hearing health monitoring, adaptive noise reduction, signal processing

1. Introduction

Occupational hearing loss remains a problem, despite the efforts made by implementing hearing conservation programs in the workplace [5]. 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 laboratory due to suboptimal placement, inconsistent use and general variations in the acoustical seal over time [21]. Despite the recent development of a field attenuation measurement system for hearing protection devices [28, 2], the precise residual noise level under the hearing protector still remains unknown [18]. 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 workers' susceptibility to develop noise-induced hearing loss [10].

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 warn the worker immediately 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 would take too much time for most standard audiological tests and would interfere with the worker's work routine.

The current research undertaken collaboratively between ÉTS and UGent aims at precisely assessing real-time variation 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 otoacoustic emission (OAE) monitoring, more specifically the measurement of Distortion Product OAE (DPOAEs).

Indeed, DPOAEs offer an objective, fast and reliable way to detect early signs of noise-induced changes in hearing sensitivity [17]. When two pure tone stimuli, f_1 and f_2 with the f_2/f_1 ratio typically around 1.22, are generated through the two miniature receivers of the OAE probe, low-level cubic distortion signals (i.e. $f_{dp} = 2f_1 - f_2$) are generated by an active non-linear process inside the inner ear. These signals travel

*Corresponding author's email address: jeremie.voix@etsmtl.ca
phone number: 1-514-396-8437

Email addresses: vincent.nadon@etsmtl.ca (Vincent Nadon),
annelies.bockstael@intec.ugent.be (Annelies Bockstael),
dick.botteldooren@intec.ugent.be (Dick Botteldooren),
jean-marc.lina@etsmtl.ca (Jean-Marc Lina)

back from the inner ear to the outer ear canal where they can be recorded. If the outer hair cells inside 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) [19], proper recording is very vulnerable to interfering background noise [25].

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. Nevertheless, no commercial system currently on the market can continuously monitor DPOAEs in a given individual, in field conditions, because the DPOAE signal is disturbed by the background noise.

Proposed Approach

To limit the disturbance of background noises, there are software solutions like noise rejection methods and Adaptive Noise Reduction (ANR) algorithms, but also hardware solutions like the improvement of the electromagnetic shielding of the probe and passive noise reduction provided by the probe eartip.

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 [3], 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. In the case of software solutions, the standard noise rejection and time averaging techniques can improve the signal-to-noise ratio (SNR) in case of limited disturbance [8], but this has shown to be insufficient in more realistic occupational noise settings [4]. 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 [8].

In response to the problems encountered with averaging methods, several ANR techniques have been studied [8, 14, 16, 24, 23]. Delgado's technique [8] 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. The ANR algorithm was proven to increase the SNR on the whole frequency spectrum while reducing the test time needed by normal time averaging methods.

In the current paper, the ANR algorithm, inspired by the work found in [8], consists 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 ipsilateral OAE probe. For the ANR algorithm, two new techniques are used in the following study: 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 ANR algorithm and a normalized version of the Least-mean square (LMS) algorithm is used to again improve signal denoising.

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 stimuli and DPOAE signals are not an integer multiple of the frequency resolution (Δf) [15]. To assess this problem Ziarani [29] proposed a method to extract nonstationary sinusoids with a non FFT based algorithm. In the current study a similar approach is used, with a temporal modulation that extracts the DPOAE signal magnitude. The combination of the ANR algorithm with the DPOAE signal extraction technique introduced in the following paper is new compared to actual DPOAE processing methods.

The major goal of this study is to develop 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. Signal extraction is validated with human subjects by comparing DPOAE results measured in quiet conditions with a commercially available clinical OAE system, the Echoport ILO292 from Otodynamics [22]. The noise reduction algorithm is tested in laboratory conditions with background noise fragments between 65 and 75 dB(A).

This paper starts with a description of the newly designed DPOAE probe (Section 2.1) and the calibration procedure for the receivers and microphones (Section 2.2). The adaptive noise reduction scheme used in this study is explained in Section 2.3. A detailed experimental procedure used for the tests is then presented (Section 2.4). Results of the comparison between the proposed system and clinical system and the tests conducted in suboptimal conditions with the ANR algorithm are then outlined (Section 3). Analysis of the results is presented in Section 4 followed by the conclusions of the study.

2. Method

2.1. Measurement Hardware Description

In order to measure a DPOAE response, an earpiece-embedded OAE probe was designed from scratch using Sonomax's earpiece body. An electronic system known as the Auditory Research Platform (ARP) [7] was developed within the Sonomax-ETS Industrial Chair in In-Ear Technologies (CRITIAS) and for this paper, it was used to send the stimuli. A computer based data acquisition system was used to capture the signals from the microphones inside the probes.

2.1.1. Earpiece-embedded OAE probe

For the prototype earpiece, two high-quality miniature balanced armature receivers were used in order to send the two pure-tone stimuli without any sound distortion. One miniature microphone was placed towards the ear canal in order to measure the otoacoustic emission response and physiological noise and a miniature microphone was placed on the outside of the earpiece to measure the external background noise (see Figure 1). Two earpieces were connected to a signal conditioning box to split the signal between the Auditory Research Platform (ARP) circuit (presented in Section 2.1.2 and shown in Figure 2) and the Data Acquisition cards. This box included a passive high-pass filter for each channel to make sure most of the undesirable signal was filtered beforehand.

In the design of the earpiece, special attention was given to the tubing guiding the acoustical signal from the receivers/microphone to/from the eartip fitting ring. This way, acoustical crosstalk was eliminated as much as possible [26] and background noise in the internal microphone was reduced due to the acoustical isolation. Roll down foam eartips were inserted onto the two probes in order to seal the ear canal for proper DPOAE measurements and also to protect the human subject when measuring in higher levels of background noise.

When tests with human subjects were carried out, the left earpiece was used to measure the DPOAE in the left ear with the in-ear microphone (IEM) and capture the external noise with the outer ear microphone (OEM). Meanwhile, the right earpiece was used to capture the physiological noise inside the right ear canal. It is assumed that the physiological noise is similar for both ear canals and since the right IEM does not capture the DPOAE responses evoked at the left ear but only noise in the ear canal, it can serve as a noise reference [8].

2.1.2. Auditory Research Platform (ARP)

In order to send the stimuli to the ear, the ARP [7] was connected via USB port to a laptop PC controlling the DSPs inside the ARP system during tests. The system was tuned to send the stimuli with a rise and fall time of 200 ms to present a clickless signal to the human ear, thus enhancing the quality of the otoacoustic response measurement by reducing wide-band noise generated by the transient nature of the click. According to recommendations of previous studies [8], the continuous stimulation period should be longer than a few hundred milliseconds in order to provide enough time for the ANR algorithm to converge, thus giving maximum performance for the algorithm. The duration of the plateau of steady stimulation of the prototype was fixed to 1.4 seconds, which is similar to the clinical reference system used in this study. With this duration, sufficient samples were collected within a reasonable test time-frame.

The 22 combinations of the two stimulus tones were generated starting with the highest frequency stimulus (e.g.: $f_2 = 6169$ Hz) and ending with the lowest frequency stimuli (e.g.: $f_2 = 1000$ Hz). This test sequence can be repeated in several loops and results can be averaged across loops. In this project up-to four loops were recorded in noisy conditions to enable robust averaging. The algorithm sending the stimuli

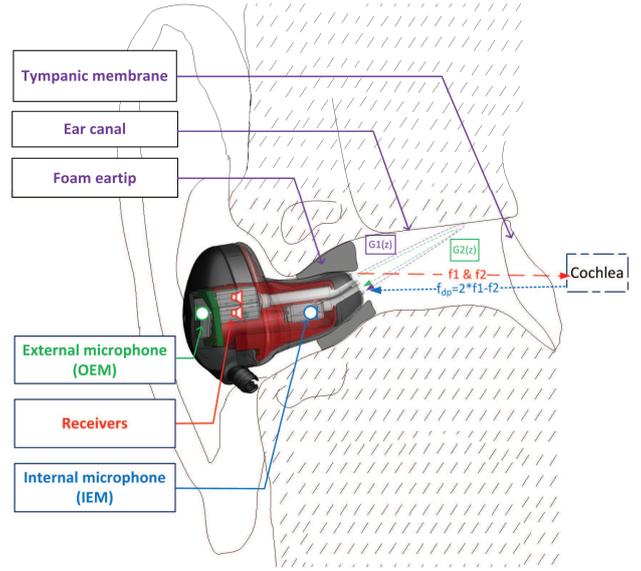


Figure 1: Earpiece-embedded OAE probe. $G_1(z)$ and $G_2(z)$ represent the acoustic path between the internal microphone and each miniature receiver. Receiver R1 generates the pure-tone stimulus with frequency f_1 and receiver R2 the second pure-tone stimulus f_2 .

was implemented in the digital signal processors (DSP) of the ARP and was optimized in assembly language in order to fit in the small memory space available. The ARP was also used to power the miniature microphones inside the earpieces with its own internal battery. The sampling frequency of the DSPs in the ARP was 48 kHz.

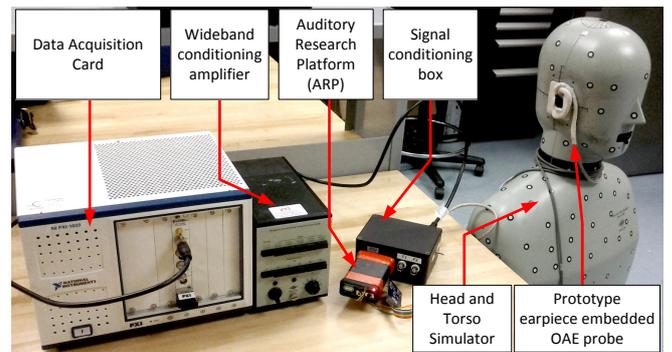


Figure 2: DPOAE data acquisition system (minimal setup shown): The Auditory Research Platform (ARP) system is connected to the signal conditioning box providing the signals to the data acquisition card and into which the prototype earpieces-embedded OAE probes are connected.

2.1.3. Computer and data acquisition card

A National Instruments (NI) NIDAQ PXI-1033 chassis (made in the USA) containing a NI PXI-4461 was used for the calibration process. The first input of the NI PXI-4461 was the

electric loopback of the signal connected to the tested receiver and the second was the amplified IEM signal of the tested earpiece.

For the acquisition of the four microphone signals (two for the left and two for the right earpieces) a NI PXI-4462 card with a sampling frequency of 48 kHz and 24 bits resolution was used to be able to synchronize all microphones easily for best performance of the ANR algorithm. The PXI-1033 was connected to a desktop computer equipped with MATLAB© version R2012b to capture and post process the signals. The input range was adjusted in order to prevent input saturation. Each input type was set to "PseudoDifferential" to reduce the AC (60 Hz) noise in the signal. A wideband conditioning amplifier (Brüel & Kjør type 2638 made in Denmark) was used for the amplification of the tested ear IEM signal only (see Figure 2), a gain of 30 dB with a linear filter was set. The signal conditioning box shown in Figure 2 included a high-pass filter for each microphone channel to reject low frequency noise and the DC signal from the microphone power supply. The purpose of the box was to connect the signals from the probes' microphones to the data acquisition card and also to be able to generate/record signals of the receivers.

2.2. System Calibration

The aim of the calibration process was to (1) calibrate all microphones for correct sound pressure level (SPL) measurement inside (IEM) and outside (OEM) the ear canal and (2) calibrate the two OAE receivers separately to set correct sound pressure levels for the pure-tone stimuli. The whole procedure consisted of two different steps described below; one to be conducted once for each new probe using a Head and Torso Simulator (HATS), henceforth called the Laboratory Calibration, and one to be conducted for each new test subject at the beginning of each measurement sequence, i.e. the Calibration with Human Subjects (see Section 2.2.2).

2.2.1. Laboratory calibration of OAE probe

The following step was carried out once for each of the two OAE probes and should only be repeated in case of malfunction. In this experiment, the Brüel & Kjør type 4128C HATS (made in Denmark) was used.

2.2.1.1. Microphone electroacoustic gain adjustment.

In order to set the electroacoustic conversion gain of the whole signal chain in Matlab, including all components and data acquisition cards, the microphone of the HATS was calibrated with a 1000 Hz pure tone of 94 dB. Based on the specifications of the HATS microphone only, 94 dB was expected to correspond approximately to 120 mV.

2.2.1.2. Initial receiver gain adjustments.

To set the correct level for both OAE receivers, the output of each separate receiver was measured with the HATS microphone. The ARP was set so that for a 1000 Hz pure tone, the HATS measured 65 dB for the R1 receiver sending the stimulus tone with frequency f_1 and 55 dB for the R2 receiver sending the second primary tone f_2 .

2.2.1.3. In-ear microphones calibration.

A white noise signal was generated by one of the OAE probe receivers to calibrate the OAE probe internal microphone and a transfer function was measured between this microphone and the HATS microphone to establish the gain adjustment necessary to match the sound pressure level at the eardrum.

Given the limited dimensions of the OAE probe and HATS ear canal, resonances were expected in the frequency range above 2000 Hz. Hence, for frequencies below 2 kHz the transfer function was expected to be nearly 0 dB due to slight differences in hardware (microphone sensitivity), signal amplification and/or signal processing.

2.2.1.4. Outer ear microphones calibration.

To calibrate each external microphone (OEM) the B&K type 4231 calibrator (made in Denmark) was set at 94 dB with a frequency of 1000 Hz. Unlike the left IEM, both OEM's were calibrated with the calibrator since these signals were not amplified by the B&K signal conditioner. During the calibration procedure, the electroacoustic gain adjustment was calculated in Matlab. Afterwards, the calibrated level was compared to the reference level of the IEM as a sanity check, since this microphone was thoroughly calibrated in previous steps.

2.2.1.5. Stimuli calibration.

In order to send the appropriate stimuli sound pressure level for each pure tone frequency, the transfer function between the receiver and the tested ear's IEM (TF3) was established. The receiver presented white noise generated by the ARP, with the overall gain adjusted by the calibration step described in Section 2.2.1.2. Figure 3 shows the schematic drawing of the test setup and the red curve of Figure 4 shows an example of a measured transfer function between the electrical signal connected to the receiver and the acoustical signal captured by the IEM. This transfer function represents the receiver's frequency response since it is assumed that the frequency response of the IEM remains flat across the tested frequencies. An initial gain table was then established per receiver at the discrete frequencies of the primary tones f_1 and f_2 , and at the calibrated reference frequency of 1000 Hz. This step was repeated once per receiver to be calibrated.

Note that in the made-up example in Table 1, the gain values are expressed as linear gain factors. At 1000 Hz, the gain factor should equal 1 as the level calibration is already established at 1000 Hz (Section 2.2.1.3) and no variation in sound pressure level is expected between the tested ear's IEM and the HATS microphone. For the primary tone frequencies, the inverse of TF3 gives the necessary gain factor that will be tabulated for further use in the ARP DSPs.

Receiver R1 - f_1 (Hz)	820	894	...	5057
Linear Gain	1	$\frac{1}{x_1}$		$\frac{1}{x_2}$
Receiver R2 - f_2 (Hz)	1000	1091	...	6169
Linear Gain	1	$\frac{1}{x_3}$		$\frac{1}{x_4}$

Table 1: Initial gain table for receivers.

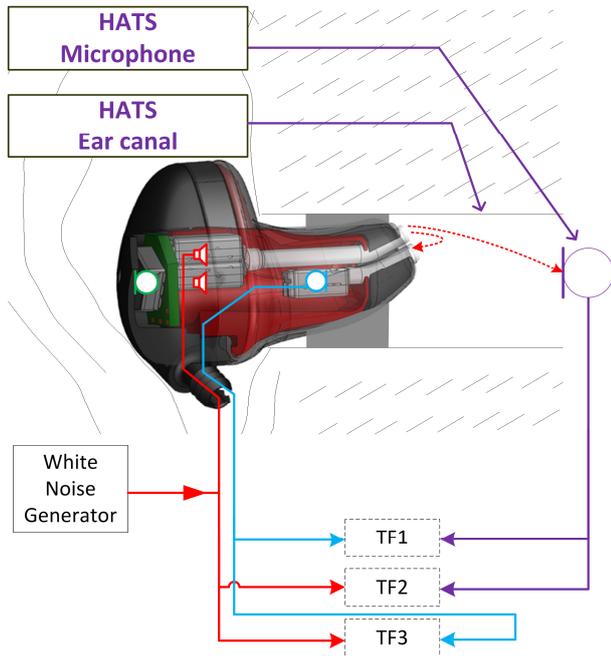


Figure 3: Schematic overview of the transfer functions measured for the calibration of the proposed system.

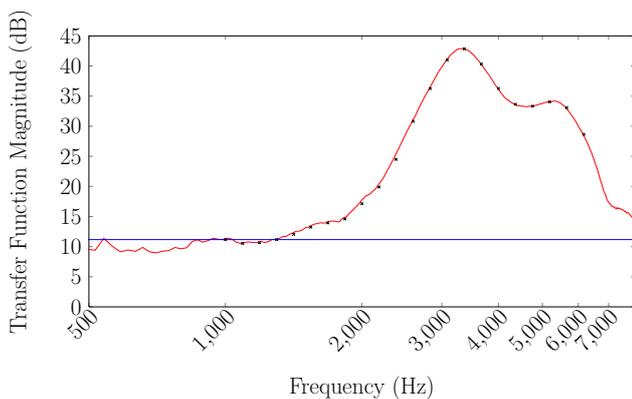


Figure 4: Transfer function between the receiver's electrical signal and the probe's internal microphone signal. The transfer function referred as TF3 represents the frequency response of the speaker to be calibrated. The horizontal blue line shown is the actual level measured at 1000 Hz. The marks represent different stimulus frequency for receiver R2. The ratio of the linear value at 1000 Hz over the linear value at the stimulus frequency gives the gain table to be applied on the receiver output.

While measuring the transfer function TF3, the transfer functions between the tested ear's IEM and the HATS microphone (TF1), as well as the transfer function between the receiver and the HATS microphone (TF2) was also measured for both receivers (see Figure 3). These transfer functions were not required for the calibration as such, but TF1 was used to give insight in the potential sound pressure level at a human eardrum based on measurements with a HATS. This was necessary to set safety limits for the OAE receiver levels on the HATS before actually working with human subjects.

2.2.1.6. Sanity check of calibration procedure.

After completing the steps above and implementing the initial gain table in the ARP (see Table 1), a final check was conducted using the HATS. Here, the primary tones were presented with the OAE receivers and the levels were recorded both at the tested ear internal microphone and the HATS microphone. The results from the tested ear IEM were used to confirm that after calibrating the sound pressure levels of the primary tones, f_1 and f_2 were indeed respectively 65 dB and 55 dB at the eardrum for every frequency. In addition, the reading of the HATS microphone in the laboratory calibration setup, confirmed that with these adjustments no excessive sound pressure level was measured at the eardrum position.

2.2.2. Calibration with human subjects

This procedure was done for each tested ear of each human subject. In theory, the gain table obtained per receiver with the HATS (Section 2.2.1.5) could also be directly applicable for human subjects — as the ear simulator of the HATS should mimic a 'standard' human ear canal — and hence this step would not be necessary. However, depending on the specific fit and the subject's ear canal shape, differences in resonance frequencies might occur from one ear to another, especially for frequencies of 5 kHz and above. Hence, it would have been unwise to use a standard gain table measured with the HATS for all subjects, thus an ear-specific gain table needed to be calculated for each new human subject. In addition, each new fit required a stimulus safety check to make sure that test subjects would not be overexposed to noise due to an error in the calibration.

2.2.2.1. Ear-specific gain table.

After placing the OAE probe in the human ear, white noise was subsequently played through each receiver (R1 and R2) separately and recorded with the IEM of the test ear. The transfer function between the tested ear's IEM and OAE receivers (i.e. TF3 in Figure 4) was checked for any abnormalities. In case of undesirable ear canal resonances, the eartip was repositioned. If a satisfying transfer function TF3 was obtained, an ear-specific gain table was produced similar to the procedure described for the HATS in Section 2.2.1.5. Finally the gain table was updated in the ARP DSP, replacing the HATS gain table with the ear-specific gain table.

2.2.2.2. Sanity check of individualized gain table.

It can be expected that in general the ear-specific gain table does not significantly differ from the HATS gain table, es-

pecially below 2 kHz where little influence of ear canal resonances/probe placement are to be seen. Hence, the linear gain at 1000 Hz should again be close to 1. As a rule of thumb, the calibration was accepted if the ear-specific gain for frequencies near 1000 Hz lied within a 0.7 – 1.4 range, i.e. allowing a 3 dB tolerance on the calibration based on empirical observations.

In order to make sure that the eardrum was not exposed to an excessive sound pressure level, the receiver output levels were limited to a linear gain of 0.3 (-10.46 dB) for the frequencies amplified by the earcanal and/or receiver (2 kHz to 7 kHz region). The linear gain was limited to 2 (6 dB) for the rest of the stimuli frequencies. Therefore, stimuli sound pressure levels were limited under 75 dB(SPL).

2.3. Adaptive Noise Reduction algorithm

In order to reduce the noise measured in the DPOAE response in noisy test conditions, the ANR algorithm was developed and is schematically presented in Figure 5. A band-pass filter was used to filter out the stimuli signal from the tested ear's IEM to help the adaptive filters converge on the DPOAE signal. Note that the same filter was also applied on all other microphones in order to keep the same noise disturbance reference.

The NLMS adaptive filter in Stage 1 modeled the transfer function between the tested ear IEM $d_1(n)$ and the contralateral IEM $x_1(n)$. The output of this filter $y_1(n)$ was then subtracted from the desired signal input $d_1(n)$. The error signal $e_1(n)$ was used to correct the adaptive filter's coefficients in order to model the physiological noise disturbance in the tested ear's IEM accurately. In the 2nd adaptive noise reduction stage, $\hat{H}(z)$ was calculated off-line. It modeled the earplug transfer function ($H(z)$). The adaptive filter was used to compensate for slight variations in the fixed $\hat{H}(z)$ estimated transfer function over time due to variations in earplug seal. The input of the fixed filter $x'_2(n)$ and the adaptive filter $x_2(n)$ was the band-pass filtered OEM signal. Each output $y'_2(n)$ and $y_2(n)$ were then subtracted from the tested ear's IEM to remove the noise disturbance.

The step size used to adjust the coefficients in the adaptive filters to process the data was set manually per subject within a range of $0.01 \leq \mu \leq 0.5$ for the first stage of the proposed ANR algorithm (see Figure 5) and the second stage was kept at $\mu = 0.01$. Only when more effect of the ANR algorithm was necessary for extreme cases, the step size was then increased up to $\mu = 0.5$. In general, a small step size will make the adaptive filter converge slowly, but the transfer function will be more precise, thus less error will be 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 was limited, especially for the second stage of the ANR algorithm, in order to not overcompensate the DPOAE response. [8] indicates that this type of overcompensation decreases the magnitude of the DPOAE response itself.

2.4. Experimental procedure

2.4.1. Background noise conditions

The current experiment aims to test the performance of the ANR algorithm when DPOAE responses are measured in elevated levels of background noise. In this study, measurements were carried out in a double-wall chamber where background noise is played through four loudspeakers. To create realistic measurement settings, an industrial noise fragment with a crest factor of 18.82 dB, dynamic range of 78.27 dB, kurtosis of 3.87 and C-A weighted signal power ratio of 2.69 dB was selected from the NOISEX database [6]. This fragment was played at three different levels: 65 dB(A), 70 dB(A) and 75 dB(A), noise spectral content is shown in Figure 6. As a reference condition, one white noise fragment was played at 70 dB(A). All fragments were played through the desktop PC soundcard, the sound pressure levels were set at the subject's position (without any subject present) using a Brüel & Kjær type 4189 free-field microphone (made in Denmark) and Nelson Acoustic Trident mX v6.8.0 software for processing.

2.4.2. Selected Test Subjects

In order to validate the proposed system, test subjects were selected based on an otoscopy test followed by tonal audiometry and tympanometry. Tonal audiometry was carried out in a double-wall chamber, meeting ANSI S3.1 [1] and ISO 8253-1 [11] standards for audiometry, using a clinical Interacoustics AC40 audiometer (made in Denmark). The subject's tonal hearing threshold should not exceed 25 dB HL at the octave band frequencies from 250 Hz to 8000 Hz. In addition, middle ear pressure, static acoustic admittance and the ear canal volume were assessed with standard clinical 226 Hz-tympanometry using a Madsen Zodiac 901 tympanometer (made in Denmark).

Before tests with the newly developed system could take place, baseline DPOAE responses were assessed with the clinical ILO292 DPEchoport system from Otodynamics (made in the United Kingdom), connected to a laptop PC with ILO v6 software. A standard clinical protocol was used where two primary tones f_1 and f_2 were presented simultaneously. The second primary tone frequencies ranged from 1000 Hz to 6169 Hz with eight points per octave, with an f_2/f_1 frequency ratio of 1.22. Stimulus levels for f_1 and f_2 were set respectively at 65 dB and 55 dB and a noise artifact rejection level of 8 mPa was used.

Measurements with the clinical system were stopped after the complete frequency range had been looped twice from the highest to the lowest frequencies. Measurements with the developed system were stopped after two loops in quiet conditions and four loops in noisy conditions. Afterwards, the signal recorded in noise was split in two fragments of two loops for averaging and test-retest purposes. Through all the experiments with the developed system, the DPOAE probe was kept in the subject's ear.

A total of 24 human subjects were selected, 11 were tested in one visit and 13 were tested in two visits (one for the ILO system measures and another for the proposed system measures in quiet and noisy test conditions). For the tests, 3 females were

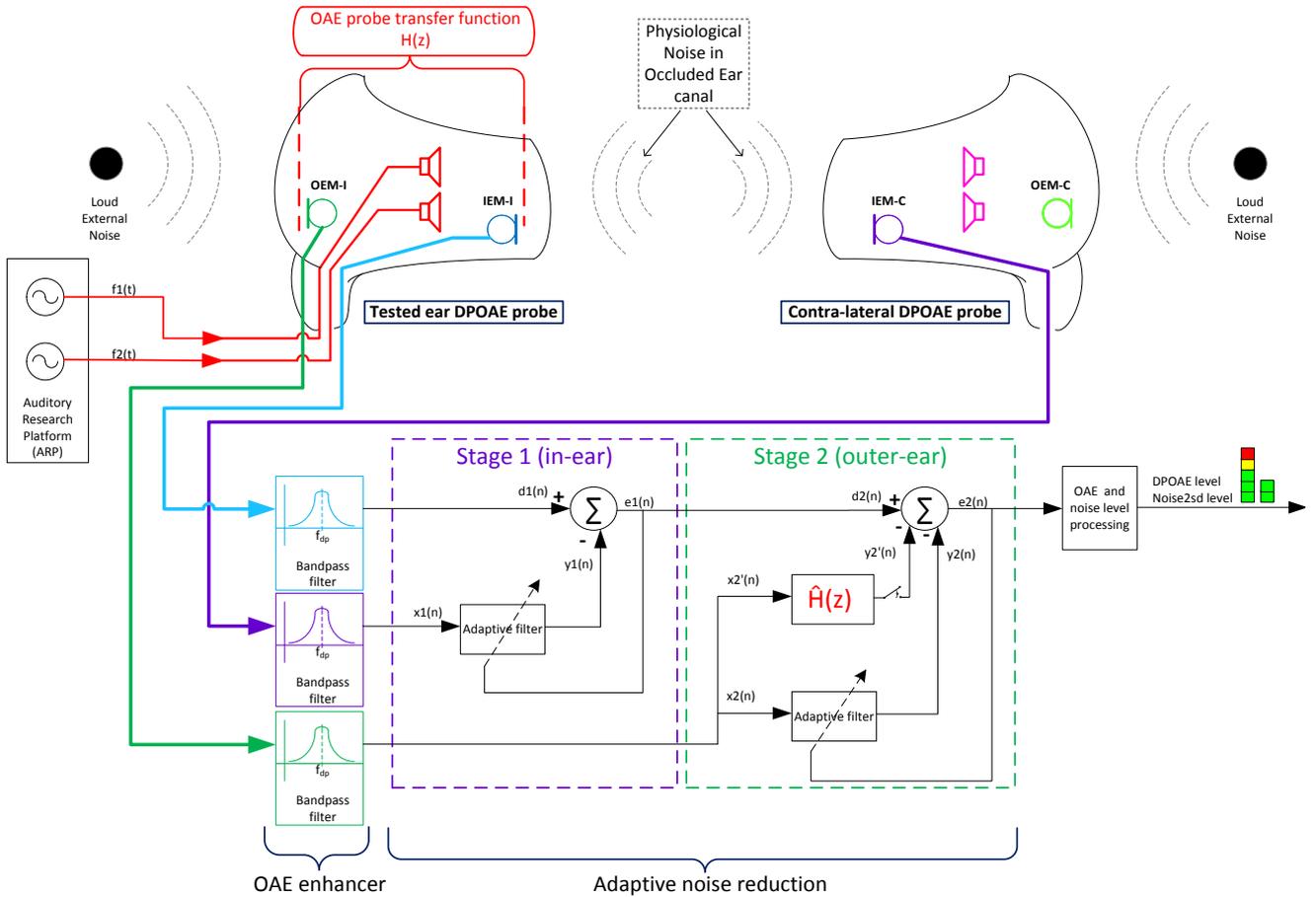


Figure 5: Block diagram of the proposed Adaptive Noise Reduction algorithm

selected versus 21 males, they were between 22 and 37 years old, the average age being 26 years old. After noise floor measurements, a hardware measurement limitation of the newly developed system was clearly noticed. To make sure that this limitation would not affect the quality of the presented results, only test subjects with DPOAE signals emerging from the noise floor – i.e. stronger than 6 dB(SPL) – were kept for further analysis (8 subjects).

2.4.3. Measurement Protocol

Subsequent to the selection test, DPOAEs were measured using the newly developed earpiece-embedded OAE probe in five different test conditions: (1) without additional background noise, in industrial noise of respectively (2) 65 dB(A), (3) 70 dB(A), (4) 75 dB(A), and (5) in white noise at 70 dB(A).

The order of these conditions was randomized over subjects and per gender. Per subject, only the left ear was tested for DPOAEs, hence the IEM of the right ear was used to assess the physiological noise. While presenting the external noise, both the test ear and the contralateral ear were occluded by the OAE probes so that even for higher levels of background noise, no threshold shifts were induced.

3. Results

In this section, results obtained with the proposed system in quiet test conditions are compared with the reference ILO system. Afterwards, results with the proposed system in noisy test conditions are presented and compared to quiet test conditions.

3.1. System comparison in quiet test conditions

To validate the proposed system, a paired comparison was conducted between DPOAE results from the proposed system and the clinical system, both obtained in quiet test conditions.

Figure 7 represents a typical subject's OAE response averaged over two recorded loops. It can be seen that the proposed systems DPOAE level follows the same trend as the ILO system over different test frequencies and for both systems the DPOAE signals clearly emerge when compared to Noise2sd, being the noise mean plus two standard deviations.

Considering the data from all the included subjects, Figure 8 shows that the proposed system has a somewhat more limited dynamic range than the clinical system and especially around 4000 Hz and to a lesser extent around 2000 Hz, the DPOAE amplitude from the proposed system are systematically somewhat lower. The systematic amplitude difference between the proposed and the clinical system is confirmed by a non-parametric Wilcoxon signed rank for paired data ($p < 0.001$).

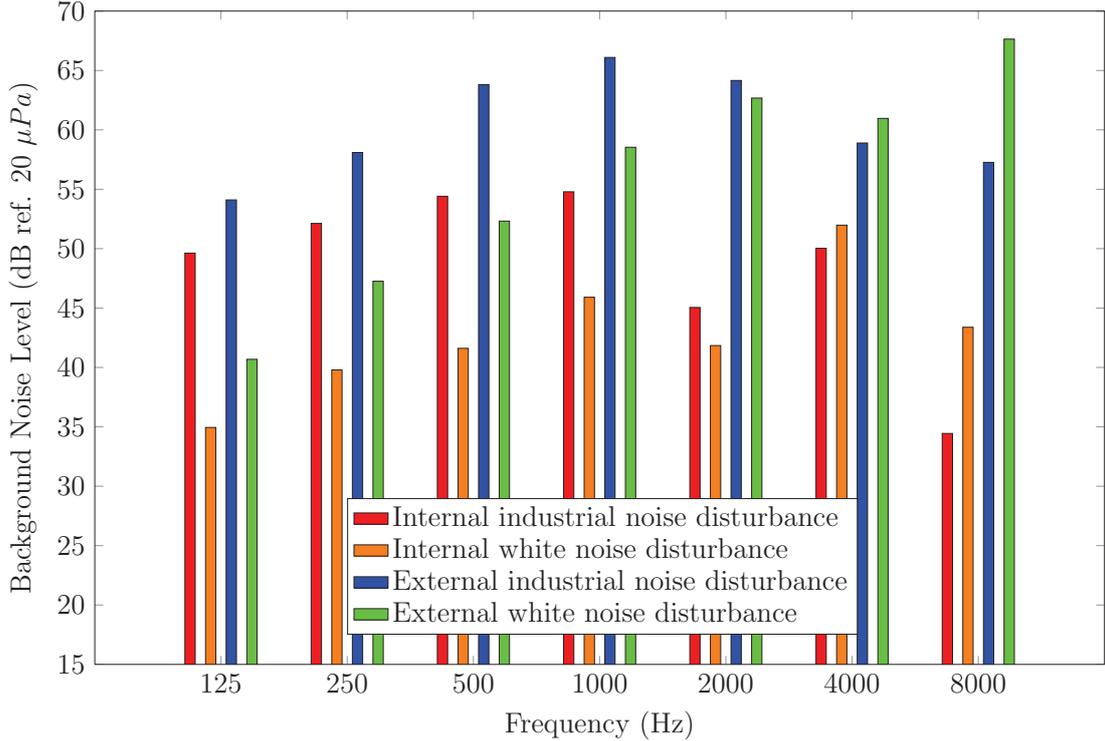


Figure 6: External microphone and internal microphone background noise disturbance level. The level of industrial noise in a first sound capture and white noise in a second sound capture in dB(SPL) are shown for the external microphone (70 dB(A) noise disturbance) and internal microphone measured per octave bands on a typical test subject while testing.

In terms of clinical relevance however, third quartile values calculated per frequency confirms that, except around 4000 Hz, clinical test-retest variability [13] is respected for the majority of observations. As discussed further, the obtained DPOAE amplitudes between both systems are therefore considered to be in sufficient agreement for this project’s purpose.

3.2. Otoacoustic emission signal extraction in noisy test conditions using Adaptive Noise Reduction

In noisy test conditions, DPOAE signals and noise are obtained by averaging the first two, out of four, recorded loops as this was shown to be sufficient. The performance of the ANR algorithm on both DPOAE signal and DPOAE noise is statistically assessed. For signal and noise separately, changes in amplitude are compared for the different combinations of background noise and ANR algorithm (e.g. industrial noise at 65 dB(A) with ANR on, industrial noise at 65 dB(A) with ANR off and so on) using a non-parametric Kruskal-Wallis rank sum test and post-hoc pairwise Wilcoxon rank sum test with Bonferroni correction.

For the DPOAE signal, both visual inspection (see Figure 9a) and statistical analysis show that the DPOAE signal amplitude varies significantly ($\alpha = 0.05$). When the ANR is switched off, DPOAE signal amplitude increases significantly ($\alpha = 0.05$) with an increased background noise level. If the ANR is switched on, the DPOAE signal amplitude is no longer systematically affected by the background noise ($\alpha = 0.05$).

As can be seen from Figure 9b, the DPOAE noise level with ANR off clearly varies for the different test conditions ($p < 0.0001$). For all conditions of background noise, the DPOAE noise levels are significantly lower with ANR on ($\alpha = 0.05$), as for ANR off, the DPOAE noise levels systematically increase with higher levels of background noise ($\alpha = 0.05$). In contrast, no significant difference in DPOAE noise level is found between test conditions with the ANR algorithm on ($\alpha = 0.05$).

According to the frequency spectrum however, even with ANR on, the DPOAE noise amplitudes still follow somewhat the spectral characteristics of the external background noise. This trend suggests that there is still somewhat room for further noise reduction with a fine-tuned ANR algorithm, although it must be said that for the largest part of the frequency range DPOAE noise amplitudes in noise lie already close to the amplitude obtained in optimal quiet conditions.

In addition to signal processing improvement, passive attenuation could also be further improved. The foam tip used for the passive sound isolation of the OAE probe did not achieve high levels of attenuation amongst test subjects. An average of 12.2 dB passive noise reduction was noted on test subjects (see Figure 10) whereas a passive noise reduction average of 30.4 dB was observed across frequencies with the probe inserted in the B&K HATS (46.3 dB insertion loss).

Moreover, the stability of the obtained DPOAE signals is assessed by comparing DPOAE signal amplitudes in noise with the baseline DPOAE signal measured in quiet reference condi-

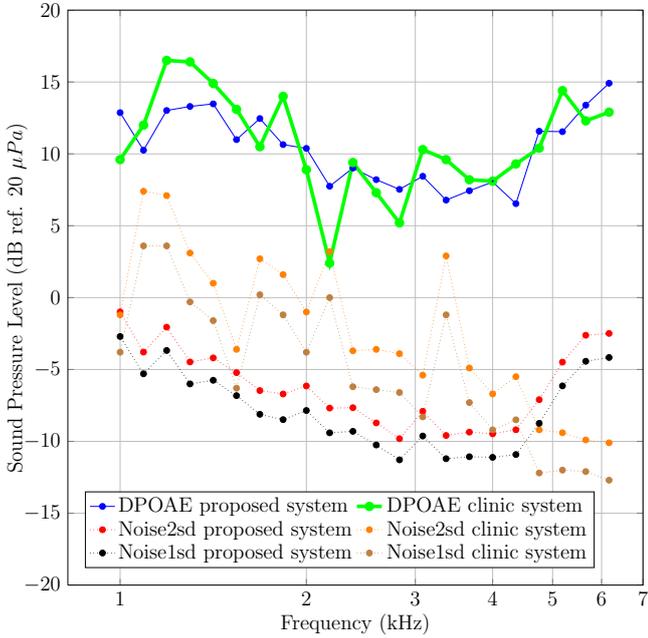


Figure 7: Typical DPgram output by the proposed system in quiet test conditions compared to the clinical system.

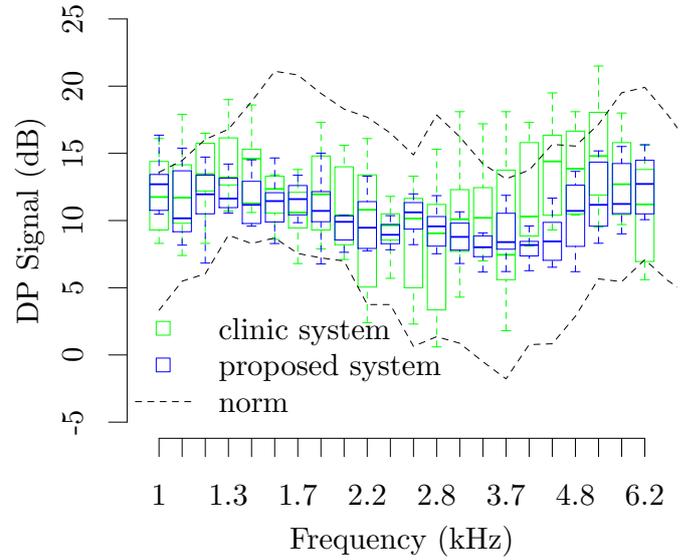
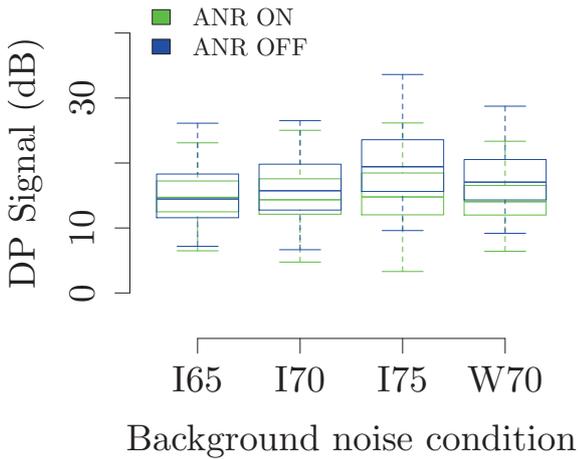
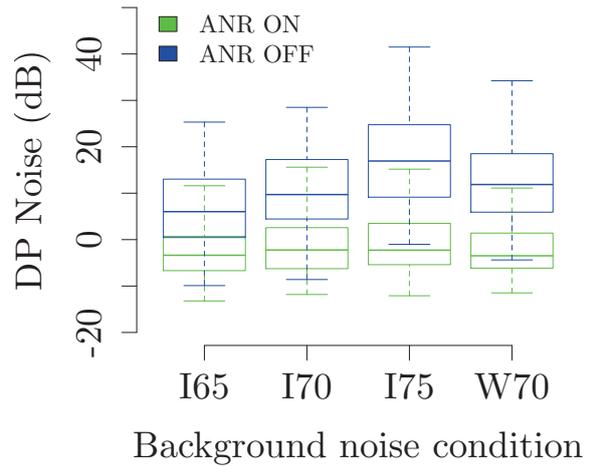


Figure 8: Box-and-whisker plot of DPOAE amplitude as a function of frequency as measured by the clinical system and the proposed system. The range of DPOAE amplitudes obtained for young, normal-hearing adults by [12] has been added as a reference ('norm').



(a) DPOAE Signal



(b) DPOAE Noise2sd

Figure 9: Box-and-whisker plots of DPOAE amplitude and DPOAE noise as a function of background noise: white noise at 70 dB(A) (W70) and industrial noise at 65 dB(A) (I65), 70 dB(A) (I70) and 75 dB(A) (I75). Calculations are made with the ANR algorithm included (ANR ON) and without the ANR algorithm (ANR OFF).

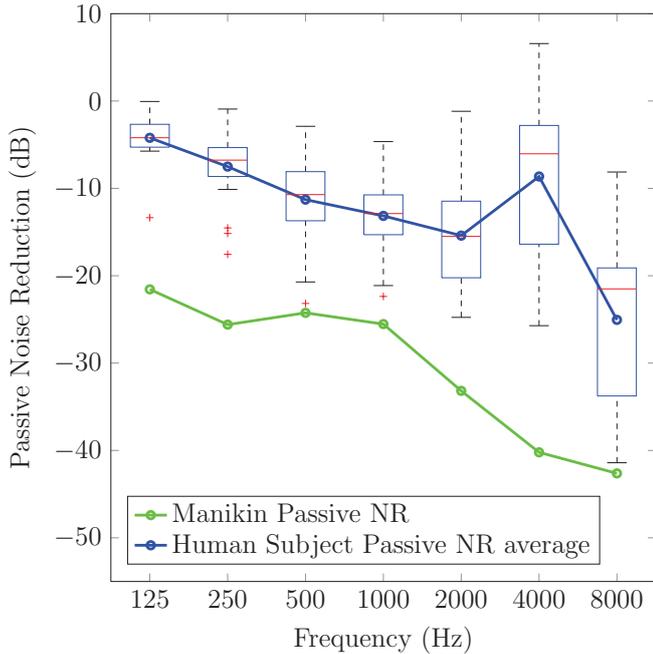


Figure 10: Intersubject Passive Noise Reduction (NR) of the roll-down foam tip used for the passive sound isolation of the developed OAE probe.

tions. A pairwise non-parametric Wilcoxon test with Bonferroni correction shows that DP amplitude does not differ significantly ($\alpha = 0.05$) in noise compared to baseline if the ANR is switched on. With the ANR off, however, the DP signal significantly deviates from the baseline conditions and this deviation increases with increasing background noise levels. The normal probability plots for the different background noise conditions (see example for 75 dB(A) in Figure 11) confirm that the DP amplitude deviation from DP baseline are centered around 0 dB difference for ANR on, and that DP amplitudes tend to be higher in background noise compared to the quiet reference conditions. An increase in DP amplitude due to measurement condition is a highly undesirable situation when monitoring cochlear changes in the workplace. An artificial increase in DPOAE signal amplitude due to measurement conditions could potentially mask a true decrease induced by cochlear damage.

Even with the ANR switch on, some observations in Figure 11 show a substantial difference with the baseline condition. The subjects with the highest DP signal deviation tend to have a somewhat lower average DP signal in quiet conditions, although more firm data is needed to reject the hypothesis that lower DPOAE baseline levels are more influenced by elevated levels of background noise. In addition, no systematic variation in DP signal in noise conditions as a function of frequency is seen here, but again, the number of subjects might be too limited to reveal this.

Based on previous work by Keppler et al.[13] the minimal detectable difference for a clinical system can be set (somewhat roughly) around 3 dB. If this limit is applied to the difference between the DPOAE amplitude in quiet and the respective con-

ditions with background noise, about 25 % of the data exceeds this limit, with slightly less for industrial noise at 65 dB(A), the least noisy condition, and slightly more at 75 dB(A), the noisiest situation.

This percentage can be reduced to 10 %-to-20%, depending on the level of background noise, when DPOAE signal amplitudes measured in industrial noise at 65 dB(A) is taken as the baseline, instead of the quiet conditions. One possible factor here is the influence of the ANR algorithm itself on the signal; in quiet conditions the algorithm has not been used whereas it is included in background noise. In the future, care will be taken that signal processing strategies are exactly the same in the baseline and different test conditions.

Finally, regardless of elevated background noise and the application of ANR algorithms, from clinical experience, DPOAE measurements are shown to exhibit some test-retest variability even without refit of the DPOAE probe [13]. To quantify this variation in the current setup, the average of the first and second loop is compared to the average of the third and fourth loops. A paired comparison between DPOAE signal in averaged loop 1-2, and loop 3-4 is made across frequencies per test condition i.e. with the ANR algorithm on and off in the different conditions of background noise. The comparison showed that for all test conditions the averaged DPOAE amplitudes do not change significantly from one pair of loops to another (paired Student's t-test, $p > 0.1$). The interquartile range of test-retest differences in DPOAE magnitude appears to be slightly higher for ANR off compared to ANR on (respectively 3.6 dB and 2.8 dB for white noise 70 dB(A) and 3.6 dB versus 2.9 dB for industrial noise at 75 dB(A)). Further data collection is needed to confirm statistically a significant effect of the ANR algorithm on test-retest repeatability.

4. Discussion

Measurements were conducted in order to validate the proposed system in quiet test conditions. The developed noise reduction algorithm — i.e. ANR — was then tested in various types of noise to evaluate its performance in suboptimal test conditions.

4.1. Quiet test conditions

As shown in Figure 8, the measured DPOAE falls within the range of DPOAE amplitudes obtained for young, normal-hearing adults. Hence, the proposed system generally leads to realistic DPOAE results falling within the range of normative data [12, 19]. While the proposed system does not exactly replicate DPOAE signals from a clinical device, results are found to be in sufficiently reasonable agreement.

Further refinements on the hardware and software should also increase the dynamic range: refinements on the hardware should reduce the electrical noise floor, hence improve the measurements at low levels; while using an enhanced scheme of the modulation algorithm should raise the upper limit of the measured DP responses.

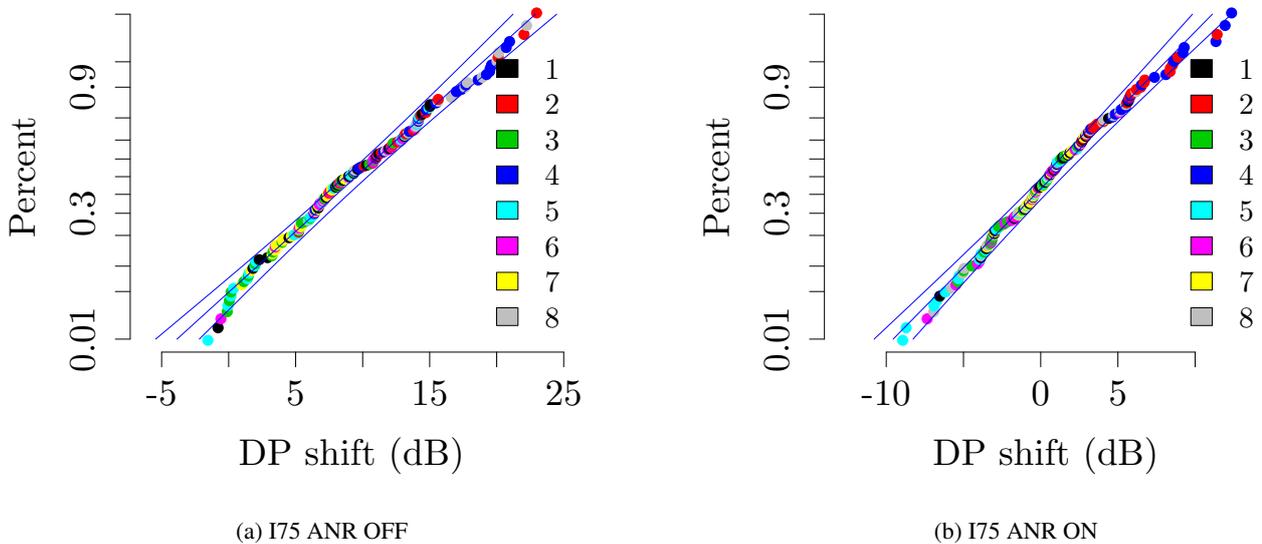


Figure 11: Normal probability plot of shift in DPOAE signal compared to baseline reference for the industrial noise background noise condition at 75 dB(A). Each subject is assigned a particular color.

4.2. Noisy test conditions

In noisy test conditions, probe fitting is very important in order to provide a good passive noise reduction therefore decreasing the influence of background noise in the DP signal. As a result, the poor passive noise reduction noted among tested human subjects (Figure 10) caused higher sensitivity to external background noise in noisy conditions for test subjects with lowest noise reduction. This effect was counterbalanced by increasing the step size of the second stage of the ANR algorithm (see Figure 5). Moreover, poor passive noise reduction could be related to poor probe fitting, hence poor passive noise reduction could engender poor stimulation levels measured at the eardrum thus leading to inaccurate DP levels for some test subjects.

In order to obtain the maximum passive attenuation from the earpiece and optimal fit for DPOAE measurements, a self-inflating custom molded earpiece, known as the SonoFit™ system [27], could be used to replace the roll down foam eartips shown in Figure 1.

Despite poor passive noise reduction, experiments with the ANR algorithm clearly showed that DPOAE signal levels obtained in noisy conditions could be brought back to the DPOAE signal levels measured in quiet conditions. As a result, the DPOAE signal level measured in noisy conditions falls within the test-retest variability observed in quiet conditions. In addition, the noise level was reduced considerably (Figure 9b), although for some frequencies the noise levels were not brought back completely to the level found in quiet conditions.

So in summary, even in absence of good passive noise reduction, the proposed ANR algorithm allows to reduce both white and industrial noise sufficiently to obtain high quality DPOAE signals even if DPOAE noise levels are not brought back ex-

actly to the levels observed in quiet. Results presented are therefore in agreement with a previous study conducted on simulated DPOAE signal extraction with noisy sound recordings [20].

In the envisioned practical use, the residual background noise level behind the OAE probe might exceed 50 dB(SPL) and, according to physioacoustic studies like the one from [9], might become an elicitor stimulus to evoke efferent activity and thereby change the OAE levels. To palliate this effect, the residual noise level will be systematically recorded together with the DPOAE levels. Therefore, when later assessing if a drift occurs in the OAE responses, only measurements made under the same noise conditions will be compared, to cancel out any physiological shift that may be introduced by the background noise itself.

5. Conclusions

A new approach for measuring distortion product otoacoustic emissions including adaptive noise reduction of both environmental noise and physiological noise was introduced. Experiments conducted in this study have shown that: 1) this approach enables the measurement of quite accurate DPOAE response in silent test conditions when compared to the clinical reference system although further hardware and software improvements may be necessary; 2) The ANR algorithm gives the opportunity to measure DP levels in noisy test conditions up to a level of 75 dB(A), which cannot be done with a standard clinical OAE measurement system even if the measurements are averaged over several loops.

Although the proposed hardware suffers from limitations, namely the poor passive attenuation of the earpiece and the high noise floor level of the measurement setup limiting the number

of selected subjects, the proposed ANR algorithm has proven its ability to denoise the OAE signals for industrial background noises up to 75 dB(A). It is foreseen that the passive isolation could be easily increased by the self-inflating custom molded earpiece of the SonoFit™ system, thus enabling DPOAE measurements in higher background noise levels.

Future work will be oriented towards the design of a new custom molded earpiece, refinement of the software and also towards hardware integration of the proposed ANR and DPOAE signal extraction algorithm in a portable DSP platform for in-field use of the DPOAE measurement system.

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