Generic acoustic test fixture for oto-acoustic emissions: principles and design

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Summary
Evoked oto-acoustic emissions (EOAEs) are routinely used to address the integrity of the auditory system at the level of the cochlea. However, standardization across different measurement devices and signal processing algorithms is still an issue, as is the variability of OAE responses depending on measurement conditions. This hampers to-date further development, interpretation, comparison, and implementation. To tackle these problems, a reference source is needed that produces realistic but time-invariant OAE responses, hence canceling out the test-retest variability seen in humans. Additionally, the acoustic test fixture has to be generic with respect to various designs of OAE equipment, meaning that all of them can operate on the simulator as they would on real subjects.In the current project, a simulator is developed to generate distortion product OAEs (DPOAEs), starting from a standard head-and-torso simulator (HATS). To operate, the OAE probe from the OAE equipment under test is placed in the ear canal of the HATS. The signals send by the OAE probe are then captured by the HATS microphone, i.e. at the location where anatomically the eardrum would be. Subsequently the primary stimuli tones are extracted in real-time by a frequency-following-filters algorithm and in response DPOAE signals are computed with a set of basic functions. The DPOAE responses are then generated by a loudspeaker mounted inside the head of the HATS and finally picked up in the outer ear canal by the OAE probe. One key aspect of the approach is working with standard hardware and transparent elementary functions so that the simulator can be easily replicated. The performance of this OAE simulator has been verified with commercially available OAE equipment, showing close agreement with normative human DPOAE data.

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1. Introduction

Oto-acoustic emissions (OAE) are low-level signals primarily originating from outer hair cell activity located in the cochlea inside the inner ear. These OAE signals propagate back from the inner ear to the external ear canal and can there be recorded by a sensitive microphone. When OAE are seen in response to an external acoustical stimulus, the term evoked OAE (EOAE) is used. Within the EOAEs, the two most commonly known types are transient evoked OAE (TEOAE) and distortion product (DPOAE) evoked by respectively click and pure tone stimuli [1].

Both TEOAE and DPOAE are frequently used to monitor the status of the cochlea and detect potential damage. OAE are routinely applied for diagnostic purposes and newborn screening [2]. In the field, they are becoming a tool for early detection of occupational and/or recreation noise-induced damage [3]. And finally, they are also applied in fundamental research studying the inner ear mechanics and physiology [4].

Generally, TEOAE appear to be more sensitive to small changes in outer hair cell functioning than DPOAE [5]. However, DPOAE are more robust, meaning that they can be measured for a wider range of hearing losses [6]. In addition, the frequency of the DPOAE response is always exactly known, facilitating accurate detection of the response, and they can be acquired for higher frequencies than TEOAE [7]. For these reasons, this project will further focus on DPOAE.

DPOAE are evoked by two pure tones of different frequencies, the so-called primary frequencies f_1 and f_2 with $f_1 < f_2$, and are seen at frequencies corresponding to algebraic combinations of f_1 and f_2. Although multiple combination DPOAEs can be obtained, numerous studies and clinical practice have focused on $2f_1 - f_2$ DPOAEs since, especially in humans, it is the component of greatest amplitude over a wide range of stimulus parameters [8]. The amplitude of the DPOAE can be optimized by varying the ratio between the frequency of the primary tones. The optimal frequency depends on the frequency of the DPOAE component under study [8], in practice a ratio f_2/f_1 of 1.22 is mostly adopted to have optimal results over the frequency range under study [9]. To optimize the amplitude of the measured DPOAE response futher, the level of f_2 is set lower than the level of f_1. When DPOAE are used to detect hearing impairment, the exact levels of the stimulus levels are critical because too high or too low stimulus levels might respectively under- or overestimate hearing damage. Therefore, intermediate primary tone level combinations are chosen, with for f_2 $L_2 = 55\ dB\ SPL$ and for f_1 $L_1 = 65\ dB\ SPL$ [10].

Compared to other audiological techniques such as the well-established pure tone audiometry, OAEs are fairly new technique and although numerous studies have been done to characterize different aspects of OAE responses (see for instance [11] and [12]), currently international accepted standardized procedures such as for pure-tone audiometry is lacking.

For standard clinical protocols and procedures, the lack of standardization can be dealt with because from literature the normal configuration of DP responses are already well-studied [13]. However, when OAE are brought to less conventional yet highly relevant test situations outside clinicals practice—for instance monitoring of early occupational cochlear damage at the work floor [3]— and/or when new equipment is developed, new validation of the registered DP responses is needed. To-date, this requires repeated measurements with human test subjects. Due to intersubject variability the test groups need to be sufficiently large, imposing practical constraints on the number of measurement conditions that can be systematically compared.

To tackle these problems, a reference source is needed that produces realistic but time-invariant OAE responses, hence canceling out the test-retest variability seen in humans. Additionally, the acoustic test fixture has to be generic with respect to various designs of OAE equipment, meaning that all of them can operate on the simulator as they would on real subjects.In the current project, a simulator is developed to generate distortion product OAEs (DPOAEs), starting from a standard head-and-torso simulator (HATS).

2. OAE simulator design

2.1. Hardware

The main focus, while deciding on hardware components of the system, was on easy, flexible and above all realistic integration of three system parts: 1. commercial OAE equipment, 2. widely used acoustical measurement equipment and 3. a software model of OAE generating mechanisms. To mimic as closely as possible a human-like OAE set-up, commercially available acoustic text fixture, namely the head and torso simulator HATS Type 4128C from Brüel & Kjær with built-in ear and mouth simulators, was used. With the microphone inside the ear simulator, the stimuli sent by the OAE device are captured and further processed by the proposed software model to generate the desired DPOAE output. Finally, the acoustic DPOAE output generated by a loudspeaker mounted inside the HATS. A block diagram of the complete OAE simulator is presented on the Figure 1.

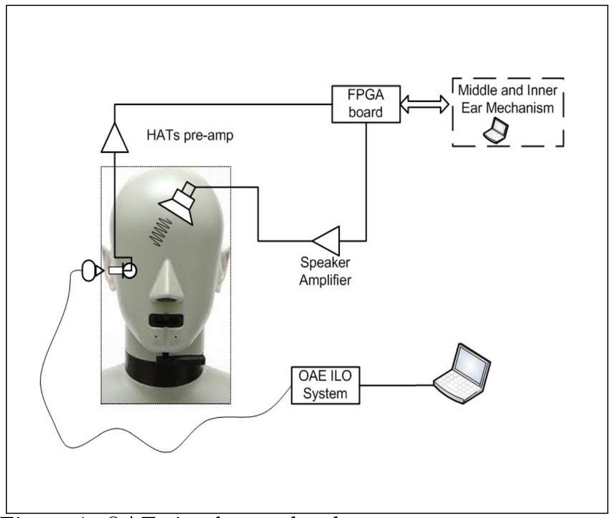

Figure 1. OAE simulator - hardware components

Taking into account that evoked OAEs by human cochlea are low level signals coupled acoustically to the OAE probe, the proposed approach allows realistic hardware integration of OAE equipment with the OAE simulator.

2.2. Software

The software part of the OAE simulator models the contribution of middle and inner ear to the OAE response. Recently, several physiologically accurate models for EOAE have been proposed, for instance by [4]. However, for the proposed test fixture we opt for a simplified design for two reasons: (1) the system should have a limited latency and thus the model should calculate fast; (2) a simplified model is easier to describe accurately and thus more suitable for standardization. A block diagram of the proposed model is shown in Figure 2.

Firstly, DC components are filtered out of the input. Further, the influence of the middle ear is known to modify OAE responses as it affects both 1. OAE stimulus signals propagating from the ear canal to the inner ear via the middle ear, and 2. OAEs generated by the cochlea, propagated back via the middle ear to the ear canal. In our model, these functions

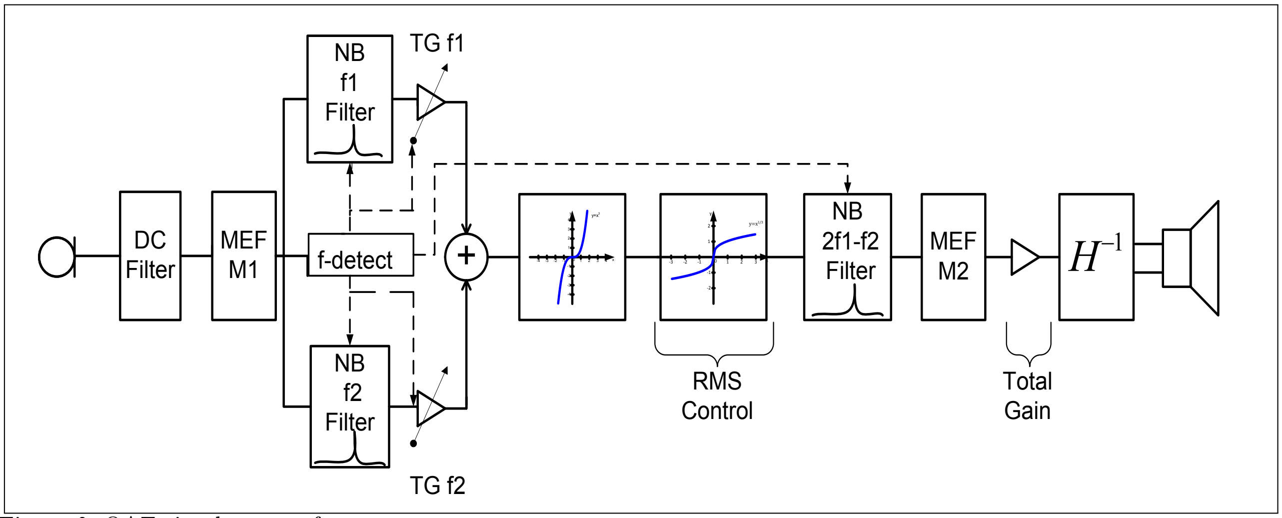

Figure 2. OAE simulator - software components

of the middle ear are represented by two FIR filters (M1,M2) of the same order $N = 1024$. The magnitude of these filters approximate average forward (M1) and backward (M2) middle ear pressure gain measured on thawed ears of human cadavers [14].

The parts of the block diagram between the middle ear filters (Figure 2) model the inner ear. As explained in the introduction, the main concern is hence to simulate the DP response at $2f_1$-f_2 for approximately $65/55dB$ primary tones excitation level and frequency ratio of approximately $f_2/f_1 = 1.22$. By introducing these constraints, precise modeling of the inner ear mechanics is not needed. Moreover, a full model would conflict with the idea of establishing an approach that is easy to replicate while reflecting in real time the fundamental cochlear mechanisms for the constrained level and frequency range. Peak filters around primary tones (f_1, f_2) are used to mimic the resonant response of the inner ear. They are designed as frequency following IIR Butterworth filters of order $N = 3$ and with Q factor of 50. Central frequency is continuously being adjusted through frequency detection procedure. This filtering is a simplified representation of tonotopic mapping within the cochlea, without replicating in detail the cochlear properties concerning equivalent rectangular bandwidth (Q_{ERB}). However, relative amplitude of the tuning curves was accounted for by introducing frequency dependent gains marked on the Figure 2 as TG f_1/ TG f_2. According to the tonotopic map model proposed in [4], linear dependency of gain with respect to the frequency tuning curve was assumed.

DPOAE response at frequency $2f_1 - f_2$ is obtained by introducing a third power nonlinear function to the model. This is the lowest power that produces these components. To obtain a realistic DPOAE response comparable to that of the average human ear, the DPOAE levels are additionally corrected in accordance the the slope of DPOAE input/output function. For the stimuli parameters of interest, this slope is reported to be $0.3dB/dB$ [15], meaning that 1 dB increase of stimuli should lead to 0.3 dB increase in DPOAE response, indicating compression mechanism of the healthy human cochlea. In the OAE simulator, compression is achieved by monitoring continuously RMS ratio of the signals before and after the third power and adjusting an adaptive gain to approximate the desired slope. This was introduced as the RMS control concept (Figure 2). Considering the level difference of 10 dB $(L_1 - L_2)$ between primary frequencies (f_1, f_2), it can be assumed that RMS ratio at input/output of the third power function (RMS_{in},RMS_{out}) is dominated by f_1 primary component. Based on this assumptions and compression level requirement, the following correction factor (CT) was finally proposed:

$$CT = \frac{RMS_{final}}{RMS_{out}} \approx \frac{A^{0.3}}{A^3} \approx A^{-2.7}$$

$$\frac{RMSout}{RMSin} \approx A^2 \tag{1}$$

$$=> CT = \left[\frac{RMS_{out}}{RMS_{in}}\right]^{-1.35}$$

where RMS_{final} is the RMS value of the signal after the correction was performed and A is the peak amplitude of the f_1 frequency component.

Before being sent to the loudspeaker, the signal is additionally filtered to compensate for the transfer path between exciter and the HATS microphone. The compensation filter is implemented as an FIR filter ($N = 1024$), identified off-line with inverse adaptive LMS technique.

Our simplified model is compared in Figure 3 to the model proposed in [4] that replicates many features of nonlinear cochlear response. Isolevel DPOAE contours for the case of $f_2 = 4000Hz$ and $f2/f1 = 1.22$

are plotted in 4dB steps as a function of L1 and L2. A good match is apparent in the level range of interest for the clinical practice (55 65 dB). Discrepancy in the other regions are to be expected given the strong simplifications made in our model leading to a lack of detail in the graph. Nevertheless, the general trend seems to be confirmed.

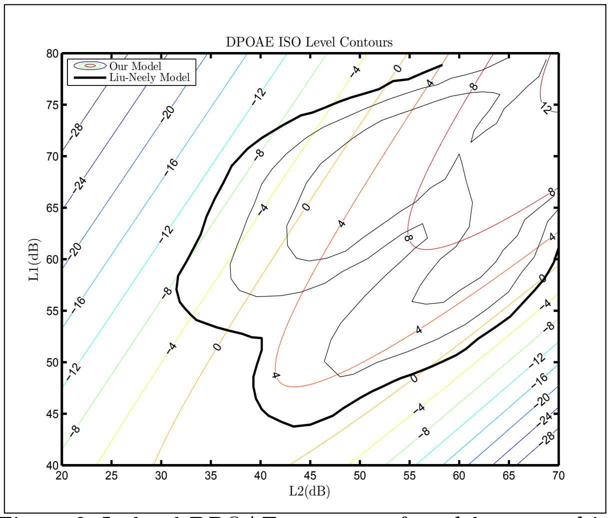

Figure 3. Isolevel DPOAE contours of model proposed in [4] (black) and our model (colored) plotted in 4 dB steps. As a reference, the contour corresponding to DPOAE 0 dB is indicated as thick black line for the model in [4], while the contours of our model are labeled.

3. OAE registration methodology

The DPOAE responses generated by the OAE simulator are registered by standard commercial equipment following the same set-up and procedure as would be done for human subjects. The clinical ILO292 DPEchoport system from Otodynamics (made in the United Kingdom) is used, connected to a laptop PC with ILO v6 software. Two primary tones f_1 and f_2 are presented simultaneously via the OAE probe placed in the ear canal of the OAE simulator. The second primary tone frequencies range 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 are set respectively at 65 dB and 55 dB and a noise artifact rejection level of 8 mPa was used.

4. Results

Figure 4 shows the DP signals generated by the OAE simulator in repeated trials. Both DP signal and noise are registered with our standard commercial OAE equipment. Over the whole frequency range, the signal-to-noise ratio of the generated DP signals compared to the registered noise clearly exceeds 0 dB and more, meaning that the DP signals could be interpreted as 'true' DP responses and not artefacts [16]. Repeatability of the measurement is very high as can be seen from the coincidence of the various measured responses. Note however that the probe was not refitted between measurements. In the mid-frequency range, the DP amplitude fits nicely the normative data obtained earlier on young normal-hearing subjects [13]. Especially in the highest frequencies, and to a lesser extent in the lower frequencies, the generated DP amplitudes drops compared to the normative data. One explanation here might be the type of OAE measurement equipment used in this particular study. Its upper and lower frequency limits are reduced compared to more recent equipment used to establish the normative data. Hence, the registration of the DP signals by the OAE equipment might be suboptimal when working on the edge of its frequency range.

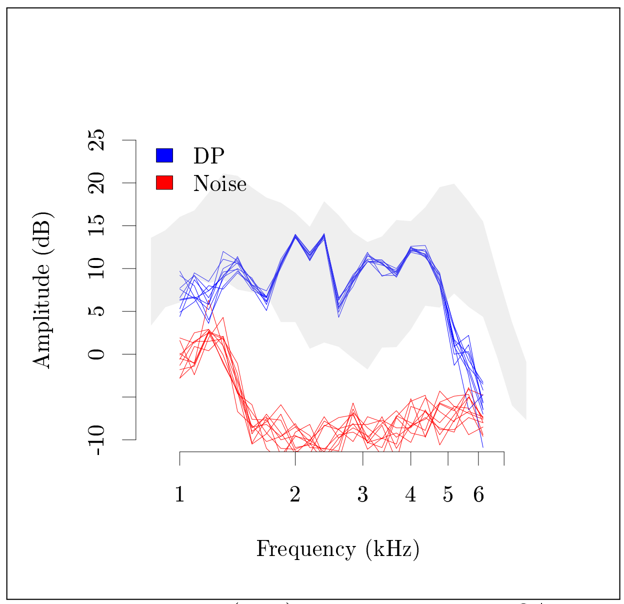

Figure 4. DP signal (blue) produced with the OAE simulator as measured by commercial OAE equipment over ten repeated measurements, depicted together with the registered noise level (red). In grey, normative DP levels are shown spanning the average DP levels obtained for normal hearing subjects plus and minus one standard deviation [13].

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