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DISTORTION-PRODUCT OTOACOUSTIC EMISSION AT LOW FREQUENCIES

– STUDIES IN HUMANS

BY
ANDERS TORNVIG CHRISTENSEN

DISSERTATION SUBMITTED 2015

AALBORG UNIVERSITY
DENMARK

Distortion-product otoacoustic emission at low frequencies

— studies in humans

Ph.D. Dissertation

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Abstract

The sensory organ of hearing, the cochlea, emits faint sound as it processes incoming sound. Measurement of such “otoacoustic emission” in the ear canal provides evidence for how the live, healthy ear works. Emissions at mid frequencies associated with speech is usually of prime interest. Low-frequency hearing has not yet been characterized by measurement of low-frequency emissions from the cochlea. Low-frequency emissions are expected to be covered in sounds of breathing, blood circulation, and so on, if they exist at all at measurable levels. The present study has measured distortion-product otoacoustic emission at unusually low $2f_1 - f_2$ frequencies, where f_1 and f_2 are the frequencies of two stimulus tones. Following a review of the literature, the first part of the study reconsiders basic choices within the standard FFT-based signal processing scheme to allow for measurements at specific f_2/f_1 ratios at low frequencies. The second part describes the design of electroacoustic instrumentation to enable measurement of emissions at subclinical frequencies. The final part relates $2f_1 - f_2$ distortion emission data around 1231 Hz to data around 246 Hz and, subsequently, presents data from 87.9, 176, and 264 Hz. The data show that the largest emission level is evoked at low frequencies when the f_2/f_1 ratio is markedly higher than at mid and higher frequencies. A relation between the optimal f_2/f_1 ratio and the equivalent rectangular bandwidth is proposed. The study shows, in essence, that the human ear emits distortion at least 1-2 octave lower in frequency than has previously been shown. The emission is promising for further exploratory and clinical assessment of cochlear activity associated with low-frequency hearing.

Resumé

Den sansende del af høreorganet, cochlea, udsender svage lyde, mens den behandler indkommende lyd. “Otoakustisk emission” målt i øregangen indikerer, hvordan det levende, sunde øre fungerer. Emissioner ved mellemfrekvenser forbundet med tale er normalt af primær interesse. Lavfrekvent hørelse er endnu ikke blevet karakteriseret ved måling af lavfrekvente emissioner fra cochlea. Sådan måling er praktisk svær på grund af anden lyd fra f.eks. åndedræt og blodcirkulation. Nærværende studie har målt emission ved usædvanligt forvrængningsfrekvenser $2f_1 - f_2$, når øret stimuleres med to toner med frekvenser f_1 og f_2. Den første del af studiet genovervejer grundlæggende valg i den standard FFT-baserede signalbehandlingsmetode for at gøre det muligt at måle ved specifikke frekvensratioer f_2/f_1 ved lave $2f_1 - f_2$ frekvenser. Anden del beskriver det elektroakustiske design af en akustisk probe instrumenteret på passende vis til lavfrekvente målinger. Sidste del af studiet relaterer emissionsdata ved en velkendt frekvens, 1231 Hz, til data ved en usædvanligt lav frekvens, 246 Hz. Endeligt præsenteres data fra stadigt lavere frekvenser, 87.9, 176 og 264 Hz. Den ratio f_2/f_1, der forårsager emission med højest niveau, er markant større ved lave ifht. højere frekvenser. Der foreslås en relation mellem den “optimale” f_2/f_1 ratio og den ækvivalent rektangulære båndbredde, velkendt indenfor psykoakustikken. Studiet viser, at menneskets øre udsender forvrængning mindst 1-2 oktaver lavere i frekvens end tidligere vist. Emissionen er lovende for videre eksplorative og kliniske studier af den specifikke cochlear aktivitet der har med lavfrekvent hørelse at gøre.

Contents

List of papers

The dissertaion is a collection of the following papers.

(A) Anders T. Christensen, Rodrigo Ordoñez and Dorte Hammershøi, "Avoiding spectral leakage in measurements of distortion-product otoacoustic emissions," *Proceedings of Forum Acusticum 2014. European Acoustics Association*, September 2014.

(B) Anders T. Christensen, Rodrigo Ordoñez and Dorte Hammershøi, "Design of an Acoustic Probe to Measure Otoacoustic Emissions Below 0.5 kHz," *Audio Engineering Society 58th International Conference: Music Induced Hearing Disorders*, June 2015.

(C) Anders T. Christensen, Rodrigo Ordoñez and Dorte Hammershøi, "Stimulus ratio dependence of low-frequency distortion-product otoacoustic emissions in humans," *Journal of the Acoustical Society of America*, vol. 137, no. 2, pp. 679–689, 2015.

    (c) Anders T. Christensen, Rodrigo Ordoñez and Dorte Hammershøi, "Erratum: Stimulus ratio dependence of low-frequency distortion-product otoacoustic emissions in humans [J. Acoust. Soc. Am. 137(2), 679 (2015)]," *Journal of the Acoustical Society of America*, accepted, in press, 2015.

(D) Anders T. Christensen, Rodrigo Ordoñez and Dorte Hammershøi, "Distortion-product otoacoustic emission measured below 300 Hz in normal-hearing human subjects," *Journal of the Association for Research in Otolaryngology*, submitted, 2015.

The following publications have also been made. They are included but not commented on further in the text of the dissertation.

(E) Anders T. Christensen, Rodrigo Ordoñez and Dorte Hammershøi, "Relating $2f_1-f_2$ distortion product otoacoustic emission and equivalent rectangular bandwidth," *Proceedings of the International Symposium on Auditory and Audiological Research*, accepted, in press, August 2015.

(F) Anders T. Christensen, James Dewey, Sumitrajit Dhar, Rodrigo Ordoñez and Dorte Hammershøi, "A Pilot Study of Phase-Evoked Acoustic Responses From the Ears of Human Subjects," *Proceedings of the 12th International Mechanics of Hearing Workshop*, June 2014.

This dissertation has been submitted for assessment in partial fulfillment of the PhD degree. The dissertation is based on the submitted or published scientific papers which are listed above. Parts of the papers are used directly or indirectly in the extended summary of the dissertation. As part of the assessment, co-author statements have been made available to the assessment committee and are also available at the Faculty. The dissertation is not in its present form acceptable for open publication but only in limited and closed circulation as copyright may not be ensured.

Preface

I was asked in 2012 before I started the PhD *"How far down in frequency can you measure this distortion emission?"* My PhD was initially meant to go in another direction, but in the first half of it, I kept getting side-tracked by my own little experiments in the lab. They had mostly to do with low frequencies and, at one point, it made most sense to pursue that direction.

I sincerely thank all the people who have been a part of my work. I wish to thank Rodrigo Ordoñez at Aalborg University, and David Purcell and Sriram Boothalingam, then both at the National Center for Audiology, ON, Canada, for helping me to initially dive into the practical and phenomenological aspects of otoacoustic emission. I wish to thank Sumitrajit Dhar and James Dewey, then both at Northwestern University, IL, USA, for having me in the lab and supporting my efforts to develop a type of measurement that might at some point aid in the separation of distortion-emission components. I wish to thank my advisors through more than three years, Dorte Hammershøi and Rodrigo Ordoñez. I believe they gave me the amount of guidance and freedom, both necessary and sufficient, to produce original research. I wish to thank Claus Vestergaard Skipper who debugged several probe-related problems with me and, when visiting his lab office, always listened to the latest uncertainties in my project. I want to thank my trusty office mate Anders Kalsgaard Møller, and finally, the subjects who patiently participated in my experimental work.

Anders Tornvig Christensen

Low-frequency distortion-product otoacoustic emission

Humans and other animals sense movement in their surroundings through hearing. Mechanical movement sets the air in motion and causes energy to propagate as sound waves that might hit the eardrum and stimulate the ear. Otoacoustic emission is in turn sound generated within the healthy ear, spontaneously, or as it actively processes incoming sound (Kemp, 1978). By measurement in the ear canal otoacoustic emission is a unique source of evidence for how the live ear works.

Low-frequency hearing has not yet been characterized by measurement of otoacoustic emission. Higher noise from blood circulation, breathing, and naturally low emission levels make the measurement increasingly difficult toward low frequencies. There has probably also been a lack of interest in trying to measure emission from the ear to characterize low-frequency hearing. Hearing sensitivity decreases rather steeply as a function of decreasing frequency, and mid-frequency content associated with speech is normally considered most important.

The following section 1 introduces briefly how hearing works and the stage of processing which is thought to generate otoacoustic emission (OAE). Section 2 reviews literature related to distortion-product otoacoustic emission (DPOAE) in humans. It focuses on what has been studied in the low end of the frequency spectrum. Section 3 summarizes the literature and section 4 presents the goal of the present study.

1 Hearing and otoacoustic emission

Sound-induced mechanical vibrations on the eardrum propagate through the middle ear into the cochlea. The cochlea houses three fluid-filled compartments separated by Reissner's and the basilar membrane in mammalian ears. These compartments extend throughout the spiral shape of the cochlea. External sounds set the membranes in a characteristic motion. Tones evoke a vibratory pattern that resembles a traveling wave, starting from the cochlear base facing the middle ear. The traveling wave then builds up displacement amplitude as it slows down approaching its place of maximum vibration (Békésy and Wever, 1960). Gradients in the physics of the basilar membrane from its base to its apex arrange the places of maximum vibration in an orderly, piano-like manner. High-frequency tones induce traveling waves that peak closer to the base than lower-frequency tones. Right after the peak the traveling waves die out and do not excite the membrane further toward the apex. Low-frequency waves pass the places associated with maximum vibration at higher frequencies. The number of cycles within the traveling wave is approximately constant across frequency, but the spread of excitation is narrower and the sharpness of the peak relatively higher for vibration at higher compared to low frequencies. Very high- and low-frequency tones do not vibrate maximally at a well-defined place.

The sound-, or motion-, sensing cells of hearing are packed in the organ of Corti which rests on the basilar membrane throughout its length (Lim, 1986). Afferent auditory nerve fibers innervate the inner hair cells almost exclusively (Spoendlin, 1972). The firing of these auditory nerve fibers is tightly related to the mechanical vibratory pattern near the places of maximum vibration, at least in the basal turns of the basilar membrane (Ruggero et al., 2000). The organ of Corti also houses outer hair cells. These are primarily innervated by efferent nerve fibers. In addition to the vibration sensitivity that inner hair cells have, the outer hair cells also have the ability to induce energy on the basilar membrane (Brownell, 1990). The outer hair cells play a crucial role in shaping the traveling waves on the basilar membrane so that they peak much more sharply at more apical places. When their function is temporarily inhibited by furosemide injected into the live mammalian cochlea, the traveling wave resembles that which can be observed in a dead cochlea (Ruggero and Rich, 1991). Furosemide-inhibited traveling waves resemble waves suppressed by the presence of another traveling wave, close in frequency and higher in amplitude (Temchin et al., 1997).

The distinguished characteristics of the live cochlea are said to be enabled by an “active process” at work in the organ of Corti, throughout the length of the basilar membrane where it rests (Davis, 1983). The active process is known mostly from experiments on isolated outer hair cells or by surgery into the basal turns of live animal cochleae. It is challenging, if not impossible, to obtain good, physiological data from more apical regions (Cooper and Rhode, 1997; Dong and Cooper, 2006). The mechanics are narrower and harder to get to without compromising the structural integrity of the cochlea and the associated lumped parameters, important in particular to low-frequency sound. An electrophysiological measure, like the compound action potential from the nerve fibers or the cochlear microphonic from the outer hair cells, is sometimes taken before and after surgery to indicate functional status quo.

Spontaneously, or as it processes incoming sound actively, the cochlea generates energy that can propagate to the ear canal. Acoustic energy in the ear canal with a cochlear origin is otoacoustic emission (OAE) (Kemp, 1978). The OAE-evoking acoustic stimulus can be isolated from the OAE itself with nonlinear or selectively suppressing stimulus techniques. In one such technique two simultaneously presented stimulus tones with frequencies f_1 and f_2 evoke distortion-product otoacoustic emission (DPOAE) at combinations of the stimulus frequencies (Kemp, 1979). This OAE provides frequency-specific data. In humans, the most prominent DPOAE occurs at the $2f_1 - f_2$ frequency and, in line with most previous DPOAE studies, it is the subject of the present study.

2 Literature review of distortion emission at low frequencies

Literature concerning $2f_1 - f_2$ DPOAE evoked in humans is reviewed. The review focuses on DPOAE evoked with f_2 frequencies below 1000 Hz in relation to that evoked with stimulus frequencies above. It gets around different stimulus paradigms applied over the years, the variety of which, to some extent, inhibits a more systematic data aggregation. The review takes note of documented challenges in making DPOAE measurements at low frequencies. There is, perhaps as a result of those challenges, no previous DPOAE data for f_2 frequencies below 500 Hz.

2.1 Early studies and basic characteristics

Kemp (1979) first reported DPOAE from human ear canals. He measured it at $2f_1 - f_2$ frequencies from 1220 to 1300 Hz, fixing both f_1 and f_2, at various narrow stimulus ratios f_2/f_1 and stimulus levels $L_1/L_2 = 60/60$ dB SPL. A highpass filter reduced the low-frequency noise from the microphone output at a rate of 12 dB/oct below 400 Hz.

Schloth (1982) presented in his PhD thesis DPOAE from three human subjects. The f_1 frequency was fixed at 1200 Hz, the stimulus ratio varied between 1.20 and 1.33 and the stimulus levels were 65/55 dB SPL. An analog bandpass filter isolated the energy at the DPOAE frequency and increasing ear-canal noise below 500 Hz was noted. The DPOAE part of the study showed DPOAE fine structure and saturation of the DPOAE level when L_2 reaches $L_1 - 5$ dB. Kummer et al. (2000); Sutton et al. (1994); Whitehead et al. (1995) have later shown that decreasing L_2 below L_1 does not affect the DPOAE level much but tends to increase reductions by hearing loss, acoustic overexposures and so on. Therefore such lowered L_2 levels are considered more relevant clinically.

Furst et al. (1988) compared various OAE measures in 11 ears of eight human subjects. Eight ears had DPOAE. The DPOAE level was measured with the stimulus ratio fixed at 1.15 for $2f_1 - f_2$ frequencies between 500 and 2500 Hz with relatively low stimulus levels 50/50 dB SPL. The DPOAE level was also measured with f_1 fixed at 1000, 1100, 1200, 1750 and 2000 Hz as the stimulus ratio varied from approximately 1.1 to 1.4. The microphone output was highpass filtered at 400 Hz. Maximum DPOAE levels were noticed for stimulus ratios near 1.2 and very low DPOAE levels for stimulus ratios greater than 1.3.

Harris et al. (1989) carried out a first systematic study of the DPOAE dependence on the stimulus ratio in 10 ears of five human subjects. The $2f_1 - f_2$ frequency was fixed at 1000, 2500 and 4000 Hz. A wide range of stimulus ratios and three different stimulus levels were tested. The microphone response (Etymotic Research Inc., ER-10) to a 65 dB SPL tone presented in the ear canal was within 5 dB from 300 to 3000 Hz and attenuated more than 20 dB at 100 Hz relative to the maximum. The DPOAE level was shown to be a bell-shaped function of the stimulus ratio with a single maximum around 1.2 and a 3-dB bandwidth of about 0.1. For stimulus levels 65/65 dB SPL the stimulus ratio evoking the largest DPOAE level, the “optimal” ratio, decreased from 1.21 at 1000 Hz to 1.17 at 4000 Hz. Later studies saw the same dependency trend in measurements at $2f_1 - f_2$ frequencies between about 600 and 5100 Hz (Abdala, 1996; Brown et al., 2000; Johnson et al., 2006; Moulin, 2000). Dreisbach and Siegel (2001) measured it up to 9000 Hz. One study by Nielsen et al. (1993) indicated differently, that the optimal ratio increased with increasing frequency for stimulus frequencies below about 1500 Hz. Other studies report consistently optimal ratio decreasing as a function of increasing frequency and also that it is consistently close to 1.22.

Gaskill and Brown (1990) also carried out a parametric but arguably less systematic DPOAE study that varied different stimulus parameters in different human subjects, 34 subjects in total. Stimulus ratio, stimulus level, level difference and frequency resolution were varied. The microphone output was level calibrated from 200 to 10000 Hz. A Knowles EA-1843 microphone with a sensitivity within 10 dB from 100 to 10000 Hz was used in a custom-built probe assembly. Highpass filtering was not mentioned. The measurements extended down to a $2f_1 - f_2$ frequency of 312 Hz, but the low-frequency end of the spectrum was not a focus of the study. The authors noted explicitly that f_1 frequencies below 1000 Hz *"were only used occassionally (...) Since low-frequency readings were particularly time consuming (due to the high noise level and the narrow bandwidths employed for measurements)."* The study demonstrated fine structure imposed on an otherwise smoothly varying, systematic dependency of the DPOAE level on the stimulus parameters. Its presence, mainly at low stimulus levels (<65/55 dB SPL), has been established in numerous later studies, for instance by He and Schmiedt (1993) and Mauermann et al. (1999). Fine structure refers specifically to an interference pattern observed across frequency with notches and soft maxima. There are typically three or four notches per third octave but their frequencies are highly individual across subjects (Reuter and Hammershøi, 2006). The implication of fine structure is that notches cancel out when measurements from many subjects are averaged. In turn, the variance across subjects is high and it is not particularly informative to relate individual DPOAE data to average DPOAE data. Under the assumption that it is two sources of DPOAE within the cochlea that interfere (Kim, 1980; Shera and Guinan, 1999; Talmadge et al., 1998, 1999), their contributions can be separated by various signal processing or suppression strategies (Brown and Kemp, 1984; Dalhoff et al., 2013; Heitmann et al., 1998; Kalluri and Shera, 2001; Martin et al., 2013; Stover et al., 1996b; Vetešník et al., 2009).

Lonsbury-Martin et al. (1990a) presented DPOAE data in relation to a number of other audiometric measures from 44 ears of 22 normal hearing human subjects. DPOAE measurements were carried out in 100-Hz steps of the $2f_1 - f_2$ frequency between 750 and 5750 Hz. The stimulus ratio was fixed at 1.21. Stimulus frequencies were selected so their pairwise geometric means (logarithmic average frequency $\sqrt{f_1 f_2}$) coincided with typical audiometric frequencies between 1000 and 8000 Hz. Minimal fine structure was seen in the measurements because high stimulus levels at 65-85 dB SPL were used, and because the low frequency resolution of 100 Hz would not expose relatively narrower fine-structure maxima or notches. A macro structure in the overall DPOAE level was seen. It has two broad maxima around the geometric-mean frequencies 1500 and 5500 Hz and a minimum between 2000 and 3000 Hz. An accompanying report of DPOAE in subjects with sensorineural hearing loss was given by Martin et al. (1990).

Probst and Hauser (1990) compared DPOAE from normal human ears to DPOAE from impaired human ears (199 ears of 101 subjects in total). Normal and impaired hearing was defined by hearing thresholds as measured in clinical audiometry. The DPOAE stimulus level was 73/67 dB HL (hearing level). DPOAE was measured at a geometric-mean frequency of 500 Hz and in half-octave steps from 1000 to 8000 Hz. As in one of very few studies, the stimulus ratio decreased with increasing geometric-mean frequency according to the recommendation by Harris et al. (1989). The stimulus ratio decreased from 1.35 at 500 Hz to 1.15 at 8000 Hz. The associated $2f_1 - f_2$ frequencies were then 280 and 6341 Hz, respectively. This is one of the lowest $2f_1 - f_2$ frequencies in the literature, aside from a few individual examples as reported by e.g. Martin et al. (2009). Only 19.5% of the 113 normal ears, however, had the 500-Hz (or really-280 Hz) DPOAE level-to-noise difference higher than 6 dB, but similar to the prevalence at the higher frequencies, 88% had the 1000-Hz (614-Hz) DPOAE. Efforts were not made to equate the level of the noise across frequency by extending the measurement duration. Dependence measurements of the DPOAE level on the stimulus level were later reported by the same authors (Hauser and Probst, 1991), but the study did not include measurements at geometric-mean frequencies below 1000 Hz, probably as a result of the low detectability in this study.

Harris (1990) measured DPOAE in 40 male subjects, 20 of whom had sensorineural hearing loss at audiometric frequencies of 3000 Hz and higher. The stimulus ratio was fixed at 1.21 for geometric-mean frequencies near half-octave steps from 1000 to 7998 Hz. DPOAE was also measured at 750 Hz but with the stimulus ratio at 1.19. The microphone output was highpass filtered at a rate of 30 dB/oct below 400 Hz. The stimulus levels were stepped from 65 dB SPL and down until the DPOAE level-to-noise difference was below 3 dB. This allowed estimation of both the normal DPOAE level, evoked by the stimulus level 65/65 dB SPL, and the DPOAE "detection threshold". The DPOAE detection threshold is the minimum stimulus level giving a DPOAE level-to-noise difference higher than 2 dB (limit varies slightly across studies), similar in principle to the audiometric threshold. Reduced maximum DPOAE levels evoked by 65/65 dB SPL significantly distinguished the two groups at 3000, 4000 and 6000 Hz. Elevated DPOAE thresholds on the other hand distinguished the two groups at 1500 Hz and above, suggesting that, to a certain extent, low-frequency emissions are affected by higher-frequency hearing loss. Such discriminating ability of DPOAE thresholds were also reported by Arnold et al. (1999) but DPOAE thresholds are time consuming to measure.

Bonfils et al. (1991) focused, perhaps as the only study in the literature, on the low-frequency end of the DPOAE spectrum. In 20 ears they measured the DPOAE at a $2f_1 - f_2$ frequency of 707.5 Hz. The stimulus ratio varied from 1.06 to 1.38 in steps of 0.02. With this configuration the stimulus frequencies varied around 1000 Hz. At each ratio the stimulus levels were stepped from 84 to 30 dB SPL in 6-dB steps. A custom-built probe assembly coupled Knowles loudspeakers and a microphone into the ear canal. There was no mention of highpass filtering. The noise level was calculated from a 50-Hz band of frequencies just below the $2f_1 - f_2$ frequency. No measurements were rejected while they were being taken, and the same amount of time, about 11 s, was spent measuring each data point, regardless of the frequency. The lowest detection thresholds were seen for stimulus ratios near 1.22 and the dependence of the DPOAE level on the stimulus level appears to have a low- and a high-level portion. For stimulus levels above 66 dB SPL, the DPOAE level increased linearly. From 40 to 60 dB SPL it tended to saturate. This saturation was taken as evidence of functioning outer hair cells. Subsequent measurements with the stimulus ratio fixed at 1.21 did not show a saturating portion for $2f_1 - f_2$ frequencies below 512.5 Hz, and from this it was suggested that *"active mechanisms are absent below 725 Hz (ie, a 512.5-Hz DPOE) in the human cochlea."* However, the higher noise level at low frequencies meant that *"no low-level portion with a plateau could be observed."*

Summary of early studies. The early DPOAE studies were exploratory and mostly carried out in small animals. There was also a variety of other types of OAE to explore in the decade after the discovery by Kemp (1978). Studies including relatively many human subjects started appearing in the late 1980s and established what DPOAE stimulus dependencies and salient characteristics were common across subjects. Difficulties in low-frequency measurements were recognized from the beginning. In addition to occasional mentioning of a high low-frequency noise floor in the literature, highpass filtering was consistently included in the recording signal chain. Highpass filtering is sensible when measurements are only made at higher frequencies and early studies do not focus on low-frequency measurements. Probst and Hauser (1990) did measure the DPOAE at a geometric-mean frequency of 500 Hz with the stimulus ratio at 1.35. They did not increase the measurement duration according to the increasing noise level toward low frequencies and, probably therefore, only 1/5 of the subjects had 500-Hz emission above the noise floor. The same problem was reported by Bonfils et al. (1991) who measured detailed DPOAE stimulus dependencies for $2f_1 - f_2$ frequencies between 342.5 and 707.5 Hz.

2.2 Clinical relevance of distortion emission

Michael P. Gorga, Stephen T. Neely and colleagues at Boys Town National Research Hospital in Omaha, NE, USA, have regularly considered the DPOAE at a relatively low f_2 frequency of 500 Hz in relation to that at higher frequencies. Their studies, reviewed below, have the stimulus ratio fixed at 1.2, which for f_2 at 500 Hz sets $2f_1 - f_2$ to 333 Hz.

Gorga et al. (1993) measured in 80 normal-hearing and 100 hearing-impaired subjects the DPOAE at f_2 frequencies between 537 and 8007 Hz (3 p/oct). The stimulus ratio was fixed at 1.2 and the stimulus levels 65 and 50 dB SPL, leveraging on the level-dependence data gathered by Gaskill and Brown (1990). It is not clear whether the microphone output was highpass filtered but the CUBDIS system was used (Allen, 1990). Efforts were not made to equate the noise level across frequency, for instance by extending the measurement duration. At f_2 frequencies below 1000 Hz, the average DPOAE level-to-noise difference was at or below zero. The authors sought to find thresholds for the DPOAE level or level-to-noise that would separate clinically normal- from impaired-hearing subjects. The DPOAE level identified hearing-impaired subjects acceptably when the hearing loss was at least 20 dB HL, but at the low frequency the prediction was as good as chance. The high low-frequency noise made the study inconclusive about the ability of the low-frequency DPOAE to indicate clinical hearing status.

Gorga et al. (1994) measured in 20 normal-hearing subjects the DPOAE at f_2 frequencies between 500 and 8000 Hz (3 p/oct). The stimulus ratio was fixed at 1.2. Stimulus level was stepped with a fixed level difference $L_1 - L_2 = 10$ dB to determine the level dependence at octave steps from 500 to 8000 Hz. Importantly, the authors used another measurement system than previously (Gorga et al., 1993). This system did not have the measurement duration fixed at 4 s but allowed an increase up to 64 s/measurement, necessary but not always sufficient at low frequencies. The noise level for each DPOAE level was also calculated in a different way. The system developed by Allen (1990) and other systems (Whitehead et al., 1994) sum the power in frequency bands near the $2f_1 - f_2$ frequency. Gorga et al. (1994) estimated instead the noise *at* the $2f_1 - f_2$ frequency in frequency spectrum of the difference signal obtained by subtracting consequtive pairs of recordings. This difference signal has signal components that remain the same across recordings cancelled out. The microphone output was highpass filtered at a rate of 12 dB/oct from an octave below the given f_2 frequency. For instance, at 500 Hz, where the DPOAE was emitted at 333 Hz, the highpass filter had its 3-dB cutoff frequency at 250 Hz. Despite these efforts implemented to cope with low-frequency noise, the conclusion was essentially same as in the previous study (Gorga et al., 1993): *“Only about 1/2 of the normal-hearing subjects had measurable DPOAEs [at 500 Hz]”* and *“First, the noise floor often remained high, even after 64 s of averaging. Second, the response amplitude was less at this frequency compared to higher frequencies, even for high primary [stimulus] levels.”*

Stover et al. (1996a) measured in 103 normal-hearing and 107 hearing-impaired subjects the DPOAE level depedence on stimulus level at nine f_2 frequencies in half-octave steps from 500 to 8000 Hz. The stimulus ratio was fixed at 1.2 and the stimulus level difference was also fixed $L_1 - L_2 = 10$ dB. The study was thus similar to the previous study (Gorga et al., 1994) and the measurement system used (Neely and Liu, 1993) was also very similar, if not identical. The study is also similar to that described by Harris (1990). The authors evaluated if DPOAE levels evoked by stimulus levels of 65/55 dB SPL would separate normal-hearing from hearing-impaired subjects, as defined by audiometric thresholds across frequency. Or alternatively, if DPOAE thresholds from the DPOAE-level growth with stimulus level would do so. For f_2 frequencies at and above 707 Hz both DPOAE level and DPOAE threshold correctly classified more than 90% of the subjects. At 500 Hz DPOAE level did not separate the groups better than chance but the DPOAE threshold distinguished more than 80% of the subjects correctly. In this latter calculation however only 65% of the subjects were included because noise at the low frequency made threshold determination unfeasible. The authors argued therefore that the low-frequency result may be misleading. In any case, DPOAE thresholds appeared to provide little, if any, improvement over the DPOAE level, attainable much faster, for f_2 frequencies at and above 707 Hz. The results in this study were more promising than those reported by Gorga et al. (1993) and Gorga et al. (1994), but the low-frequency noise was again problematic. The authors noted specifically *“The high noise floor for lower frequencies occurred across all subjects independent of hearing status and made data interpretation more complex.”* It remains curious that the ability of the DPOAE level to indicate audiometric threshold status changes as much as this study shows between the two low frequencies, from chance prediction at 500 Hz to correct prediction in more than 90% of the subjects at 707 Hz and above. A similarly discouraging result for the low-frequency DPOAE was obtained by Probst and Hauser (1990), summarized above, but they made no effort to equate the noise level across frequency.

Gorga et al. (1997) measured in 1267 ears of 806 subjects the DPOAE at similar parameters as previous studies but this, derived and later studies by Gorga, Neely and colleagues did not – not surprisingly at this point – include or consider the DPOAE evoked with f_2 frequencies below 750 Hz and in some cases below 1000 Hz.

Summary of clinical relevance. The ability of the DPOAE to reflect audiometric hearing status was evaluated by Gorga, Neely and colleagues in relatively large groups of normal-hearing and hearing-impaired human subjects. The two measures may have a related but not the same physiological origin. It is not surprising that there is not an unambiguous relation between the two. The results from the studies about the relation appeared conclusive for frequencies at and above 707 Hz but possibly "misleading" at 500 Hz. The efforts made to cope with the higher noise floor at low frequencies, mainly by extending the time spent measuring individual points, were not sufficient. DPOAE measurement with f_2 at 500 Hz appears to require different methodology, for instance by different stimulus parameters, steeper highpass filtering of the microphone output, better transducers, different rejection of "noisy" measurements, noise-floor calculation, or a refined combination of all such efforts.

2.3 Distortion emission at a low and high frequency

In later studies Gorga, Neely and colleagues take on less clinically-oriented objectives to assess low- and high-frequency "cochlear nonlinearity" and low- and high-frequency DPOAE-suppression tuning curves.

Gorga et al. (2007) measured in 103 normal-hearing subjects the DPOAE dependence on stimulus levels for f_2 frequencies 500 and 4000 Hz. The stimulus ratio was fixed at 1.22 which sets the associated $2f_1 - f_2$ frequencies to 320 and 2557 Hz. It was stated explicitly that *"0.5 kHz was chosen because it is perhaps the lowest frequency for which reliable DPOAE data could be collected, due to problems associated with the increase in noise levels as the frequency decreases."* The two frequencies were also selected to reflect apical and basal cochlear processing which, some literature suggests, works by different physiological mechanisms (Dong and Cooper, 2006; Nowotny and Gummer, 2006; Reichenbach and Hudspeth, 2010; Shera et al., 2000, 2013). Gorga et al. (2007) specified the stimulus levels in dB SL (sensation level) and not in the typical dB SPL, in an effort to equalize the "internal" stimulus to the cochlea. The low-frequency measurement was allowed to take up to 210 s/point as opposed to the earlier 64 s/point (Gorga et al., 1994; Stover et al., 1996a). For low-level stimuli such lengthy measurement durations were necessary and not always sufficient. The authors returned to the classic way of estimating the noise at the $2f_1 - f_2$ frequency from the integrated power in neighboring frequency bands (Gorga et al., 1993), and not from the power at the $2f_1 - f_2$ frequency in the difference spectrum (Gorga et al., 1994, -on). The DPOAE level was a steeper increasing function of increasing stimulus level at 4000 Hz compared to that at 500 Hz. Up until the highest stimulus levels (<30 dB SL or <50 dB SPL) it took a higher stimulus level to evoke the same DPOAE level at 500 Hz as at 4000 Hz. Noise levels at the two frequencies were rarely the same. The 500-Hz noise was usually about 10 dB higher than the 4000-Hz noise and, therefore, the results were not conclusive about the actual low-frequency DPOAE threshold (and dynamic range). The slope for instance may be shallower and extend the threshold to lower stimulus levels than at 4000 Hz. The estimated dynamic range of the DPOAE level relative to the stimulus level was about 20 dB smaller at 500 Hz than at 4000 Hz. This indicated more compression or less "amplification" at low frequencies, also seen in behavioral equal-loudness contours (Møller and Pedersen, 2004).

Gorga et al. (2008) measured in 19 normal-hearing subjects suppression of the DPOAE level for f_2 frequencies 500 and 4000 Hz by a third stimulus tone. A range of DPOAE stimulus levels specified in dB SL were tested. The tuning curves were obtained by variation of the suppressor-tone level (dB SPL), until the DPOAE level was suppressed 3 dB. Suppressor frequencies extended from -1 to 0.5 oct re f_2 at 4000 Hz and from -2 to 0.5 oct re f_2 at 500 Hz. Effectively, the suppressor tone extended down to 125 Hz. The study took 40 h/subject spread over several sessions of data collection. At 4000 Hz maximum suppression was seen when the suppressor frequency was about 0.1 above the f_2 frequency, regardless of the L_2 level. At 500 Hz the suppressor frequency giving most suppression decreased with increasing L_2 level, from 0.15 oct re f_2 at 20 dB SL to -0.1 oct re f_2 at 50 dB SL. The place on the basilar membrane of maximum DPOAE generation may shift toward the apex as the stimulus levels increase at low frequencies. It took a 20-dB higher suppressor level to suppress the 500-Hz DPOAE level maximally at low L_2 levels but the difference decreased to about 5 dB at high L_2 levels. The DPOAE-generating mechanisms in the apical and basal turns of the cochlea may be similarly active at higher stimulus levels. Off the place of maximal suppression it took markedly higher suppressor levels to suppress the DPOAE level at 4000 Hz. The active cochlear process may affect cochlear responses more broadly in the apical turns of the cochlea and more locally in the basal turns of the cochlea.

Summary of distortion at a low and high frequency. The DPOAE is more sharply tuned at the high than at the low frequency. At the frequency of maximum suppression, in comparison to the low-frequency DPOAE, the high-frequency DPOAE takes on a wider range of levels as a function of the stimulus level. At the high frequency the suppressor frequency suppressing the DPOAE level the most does not depend on the stimulus level. At the low frequency it decreases with increasing stimulus level. These differential characteristics reflect relative changes in the DPOAE-generating mechanisms and may have a relation to physiological and psychophysical tuning data (Charaziak et al., 2013; Cooper and Rhode, 1997; Jurado and Moore, 2010; Shera et al., 2013). In such comparisons, and in general, it may be of relevance that only the stimulus levels were optimized at the two frequencies. The DPOAE data were gathered with the stimulus ratio kept fixed at 1.22. This fixed stimulus ratio allows for comparison with almost all previous data across frequency, but, for fixed-level stimulus tones, maximum DPOAE levels are evoked by increasing the stimulus ratio as the stimulus frequencies decrease (Harris et al., 1989; Johnson et al., 2006).

The results by Gorga et al. (2008) are generally more conclusive at the low frequency than previous studies by Gorga and colleagues. There were not major differences in the methodology of the measurement, but much more time was spent measuring the DPOAE. In measurements of suppression tuning curves, adjacent measured points validate each other when they follow an overall trend, and the presence of adjacent measured points makes the integrity of single points less important.

2.4 Distortion emission from basal sources

Glen K. Martin and colleagues at Loma Linda University Medical Center in Loma Linda, CA, USA, have carried out a number of studies that suggested basal sources of DPOAE. “Basal” sources are supposedly increasingly relevant in low-frequency DPOAE measurements, because the basal spread of the basilar-membrane excitation by a stimulus tone increases as its frequency decreases, or as its level increases. Following a brief introduction to where in the cochlea the DPOAE is thought to arise, selected studies by Martin and colleagues are reviewed.

The DPOAE is generated by nonlinearity in the region of the basilar membrane excited by both stimulus tones simultaneously (Kim, 1980). That corresponds to the region excited by the f_2 tone. The excitations by the f_1 and f_2 tones are presumably similarly sized where the excitation by the f_2 tone is maximal. The prevailing view is therefore that the f_2-related source of DPOAE is focused in a relatively narrow region around the place of maximum excitation by the f_2 tone. From the narrow f_2-related region the generated energy is thought to propagate both basally and apically. The apical-directed energy propagates past the f_1 place toward the $2f_1 - f_2$ place of maximum excitation. Around this place the local outer hair cells amplify the vibration and locally-dense impedance pertubations cause partial reflection (Zweig and Shera, 1995). The reflection propagates basally and adds to the original basal-directed energy. The emission measured in the ear canal is then composed of contributions from two narrow regions on the basilar membrane, that related to the f_2 place and that related to the $2f_1-f_2$ place of maximum vibration. The two contributions are related because one is evoked by the other. They arise as a result of two fundamentally different mechanisms, namely nonlinear distortion near the f_2 place and linear coherent reflection near the $2f_1 - f_2$ place (Knight and Kemp, 2001; Shera and Guinan, 1999).

In addition to the f_2-related source, and its reflection from the $2f_1 - f_2$ place, there may be a basal source of DPOAE. “Basal” source does not refer to the narrow f_2-related source when the stimulus frequencies drive basal regions of the basilar membrane. It refers to a source residing in the basal skirt of excitation by the f_2 tone. The nonlinear source can be considered an f_2-related DPOAE source, and the “basal” source a basal-to-f_2 source. The spatial extent of the two sources, and whether they are distinctly separated or not, is not clearly established. The basal source is sometimes regarded a spatially distributed f_2-related source.

Arnold et al. (1999) tested if very high-frequency hearing loss affected lower-frequency DPOAE. High-frequency hearing thresholds were measured at 8 frequencies from 9000 to 20000 Hz in 50 human subjects with normal audiometric thresholds from 250 to 8000 Hz. DPOAE was also measured at 10 p/oct from 800 to 8000 Hz. The stimulus ratio was fixed at 1.22 and the stimulus levels were 75/75 dB SPL as in the reference data study for normal-hearing subjects obtained by Lonsbury-Martin et al. (1990a). Thirty-one subjects were also tested at 55 and 65 dB SPL. The average hearing threshold from 11200 to 20000 classified good and poor high-frequency hearing subjects, and these were compared statistically to the DPOAE level from 4000 to 8000 Hz. Only DPOAE levels evoked by the high stimulus level, 75/75 dB SPL, were significantly different between the groups. This may have been the case, either because fewer subjects were tested at the low level, or because basal sources of DPOAE, affected by very high-frequency hearing loss, contribute relatively more at high stimulus levels. The finding that hearing loss at much higher frequencies than the DPOAE-related frequencies can affect the DPOAE may account for the modest relation between hearing thresholds and DPOAE levels observed by Gorga et al. (1993); Stover et al. (1996a), particularly at low frequencies where the basal extent of the stimulus excitation is larger.

Martin et al. (2003) measured in 21 normal-hearing humans suppression of the DPOAE at geometric-mean frequencies 1000, 2000, 3000 and 4000 Hz by a third tone. This study extended elaborate evidence in rabbits (Martin et al., 1999). The ratio was fixed at 1.22. Three stimulus levels L_1/L_2 at 85/85, 75/75 and the more common 65/55 dB SPL were tested. The major difference between this and other DPOAE-suppression studies was the extent of the third-tone suppressor-frequency and -level variation. In all stimulus conditions the suppressor tone was swept from 250 to 9500 Hz at levels from 35 to 85 dB SPL. In agreement with previous suppression studies, notable suppression was found when the suppressor frequency was near the stimulus frequencies, but suppression and enhancement lobes were also found in a suppressor-frequency range near $2f_1$. That is significantly above f_2 where the suppressor is not expected to interfere with the f_2 vibration. Enhancement was evident at the low stimulus level, 65/55 dB SPL, for the low geometric-mean frequency tested, 1000 Hz. At higher frequencies and higher levels, enhancement was not seen, but suppression of the DPOAE was present when the interference-tone frequency was near the harmonics of the stimulus frequencies. The theory for basal-to-f_2 suppression/enhancement is compelling but too elaborate to be described in detail here (Fahey et al., 2000; Martin et al., 1999). Briefly, a harmonic of the f_1-stimulus tone, generated within the cochlea, is thought to interact with the fundamental f_2-stimulus tone. In combination, the two may produce the quadratic-difference distortion product $(2f_1) - (f_2)$ as opposed to the cubic odd-order distortion product $(2f_1 - f_2)$. Depending on the phase relationship between the resulting basal-to-f_2-related and f_2-related DPOAE sources, suppression of the basal source by an interference tone may result in a net decrease (suppression) *or* increase (enhancement) of the total DPOAE level.

Martin et al. (2009) showed, by DPOAE measurement in three human subjects for f_2 frequencies between 500 and 6000 Hz and a widely varied stimulus ratio, that the steep phase gradient, normally associated with the DPOAE contribution from the $2f_1 - f_2$ region on the basilar membrane (Shera and Guinan, 1999), could not be removed by a high-level suppressor tone at $2f_1 - f_2 - 44$ Hz. A similar result was obtained by Dhar et al. (2011). Instead, a similar suppressor placed 1/3 octave above the

f_2 frequency removed the steep phase gradient, suggesting a significant basal contribution to the DPOAE. It did not however remove fine structure from the DPOAE-level response. High-level stimuli at 75/75 dB SPL were used. The basal source appears to mainly be active at high levels, low frequencies, narrow stimulus ratios, or all, where the relative basal spread of the stimulus excitation is largest.

Martin et al. (2010) and Martin et al. (2013) found further evidence of basal sources in rabbits. Martin et al. (2010) demonstrated that the basal contribution can mask a reduction in the DPOAE level due to noise-induced hearing loss near the characteristic places of the stimuli. Once the basal contribution was suppressed by a third tone presented 1/3 octave above f_2, the DPOAE-level reduction closely resembled the noise-induced reduction in the auditory brainstem response. Notched higher-frequency losses were also better identified. Martin et al. (2013) demonstrated basal sources in time domains of measurements in rabbits.

Summary of basal sources. High-frequency hearing loss can affect DPOAE evoked by lower-frequency stimulus tones. An explanation for this observation is that DPOAE is not only generated where the f_2 excitation peaks on the basilar membrane, but also significantly basal to this place. The basal-to-f_2 region of DPOAE generation is referred to as a “basal” source of DPOAE. Its main implication for the clinical utility of DPOAE is that the DPOAE apparently does not arise in regions on the basilar membrane that are as narrow as it was thought earlier. The basal source is more evidently present when the stimulus tones are close (narrow stimulus ratio), relatively high in level and low in frequency because these stimulus conditions increase the basal spread of the basilar-membrane excitation. At the lowest stimulus frequencies tested so far, near 1000 Hz, both suppression *and* enhancement of the DPOAE level by a third tone was observed. Perhaps at even lower frequencies the effects and underlying causes of basal sources are more clearly evident.

It should be noted that a basal source of DPOAE does not contradict the existence of the f_2-related source or its reflection at the $2f_1 - f_2$ place.

The three probably contribute different amounts in response to different stimulus parameters. The f_2-related source contributes consistently to the DPOAE in the ear canal across a wide range of stimulus parameters. The reflection from the $2f_1 - f_2$ place is largest when the stimulus by the internal source(s) is small, for instance when the stimulus level is small. In turn, the basal source appears relatively large for narrow ratios, high stimulus levels and low frqeuencies. Understanding and coping with the differential presence of these sources is important for clinical utilization of DPOAE.

2.5 Other observations at low frequencies

The scope of the review is limited to basic characteristics and stimulus parameters of the $2f_1 - f_2$ DPOAE in humans. Other aspects of DPOAE at low frequencies that have been studied deserve to be mentioned.

Phase of the distortion emission. The DPOAE phase is approximately invariant as a function of frequency when standard stimulus parameters are used, that is, a stimulus ratio fixed near 1.22 and stimulus levels 65/55 dB SPL. Some phase fine structure associated with the rippled fine structure in the DPOAE level is normal. For f_2 frequencies below approximately 2000 Hz the phase is typically not invariant but decreases instead a half-to-whole cycle per octave. Sometimes the transition with frequency is rather abrupt and the change in phase slope is often referred to as a “break”. Dhar et al. (2011) studied in 10 normal hearing humans the break in phase invariance specifically for f_2 frequencies between 500 and 4000 Hz, and the systematic phase decrease with decreasing frequency extends at least to 500 Hz. The origin is not known but it may be a result of differences between apical and basal (active) cochlear mechanics (Shera et al., 2000, 2013).

Age and middle-ear influence. Lonsbury-Martin et al. (1991) compared DPOAE levels at geometric-mean frequencies between 1000 and 8000 Hz in groups of 20-, 30-, 40-, and 50-year old subjects. The latter three groups were not significantly different from one another but all were slightly different from the young 20-year old group. The difference was similar in hearing thresholds and DPOAE levels and was only apparent at frequencies above 2000 Hz. DPOAE thresholds in the 50-year old group was slightly different from the younger groups across frequencies. Abdala and Dhar (2012) compared DPOAE levels and phase in subjects ranging from premature newborns (37 wks on average) to older adults (69.2 yrs). The hearing thresholds were normal in all groups except in the oldest group. Correspondingly, there was not a clear tendency in the DPOAE level with increasing age, except older infants (6-8 mos) had markedly high DPOAE level and the oldest subjects had markedly reduced DPOAE levels across frequency, a flat reduction within few dB in comparison to the middle-aged group of subjects (48.8 yrs). Poling et al. (2014) showed a clear distinction in DPOAE levels between young and older subjects for f_2 frequencies between about 3000 and 13000. Their figures have a linear scale which makes it difficult to see clearly the low-frequency tendency with age, but it appears to be smaller than at higher frequencies. As another DPOAE measure of age, the low-frequency phase slope, and the break frequency, appears to distinguish differently aged humans (Abdala and Dhar, 2012). The changing low-frequency phase slope may be related to the maturation of the middle ear (Abdala and Keefe, 2006), and to changes in the apical turns of the cochlea with age (Abdala et al., 2011). The middle ear is generally expected to affect the lowest-and highest-frequency emissions most because changes in the lumped stiffness and mass parameters change the transmission of both stimulus and response. Studies of middle-ear effects seem limited in their extent to extreme frequencies, live humans and measures beyond tympanometry, e.g. Lonsbury-Martin et al. (1990b); Osterhammel et al. (1993); Puria (2003).

Posture influence. Voss et al. (2006) measured in seven female human subjects the effect of posture on the DPOAE level at f_2 frequencies between 750 and 3984 Hz. The measurement duration was longer at the low frequencies in an attempt to equate the noise level across frequency. The stimulus ratio was fixed at 1.2 and the stimulus level was 65/55 dB SPL. Subjects were tilted on a table to upright and horizontal positions and so that their head were below their bodies at angle of -30 and -45 degrees. Upright and horizontal postures did not change the DPOAE level. But the -30- and -45-degree postures produced highly significant reductions in the DPOAE level of more than 6 dB. The effect started at 2000 Hz and was largest at the lowest frequency, 750 Hz. The authors attributed the observed effect to a change in the intracochlear pressure, as induced by a change in the pressure of the fluids in the skull, the intracranial pressure. A more elaborate, confirming study was later carried out by Voss et al. (2010).

Distortion emission in animals. There may be examples in the literature of relatively low-frequency measurements in anesthetized animals. Meenderink et al. (2005) for instance made in five anesthetized female Northern leopard frogs measurements of DPOAE at f_1 frequencies between 213 and 2774 Hz. The stimulus ratio varied between 1.02 and 1.70, setting the lowest $2f_1 - f_2$ frequency to 63.9 Hz. They did so without special instrumentation and the frogs did not appear to have emission of significance below an $2f_1 - f_2$ frequency of 500 Hz.

Other distortion products. When the $2f_1 - f_2$ DPOAE is evoked with a stimulus ratio between 1 and 1.5, other DPOAE components may be present at lower frequencies, for instance at $f_2 - f_1$. An f_2 at 500 Hz and a stimulus ratio at 1.22 set $f_2 - f_1$ to 90 Hz. Bian and Chen (2008) measured the $f_2 - f_1$ DPOAE down to 500 Hz. The $2f_2 - f_1$ frequency, where another DPOAE resides, is always higher than the stimulus frequencies. It may therefore be present for very low stimulus frequencies. However, the electroacoustic instrumentation is designed for measurements of the $2f_1 - f_2$ component because this is most prominent in humans. Such instrumentation typically filters the microphone output at a rate of 12 dB/oct below a 3-dB cutoff frequency at 250-500 Hz.

3 Literature summary of distortion emission at low frequencies

Studies of the $2f_1 - f_2$ DPOAE in more than a handful of human subjects started appearing in the late 1980s. Its basic dependencies on the stimulus parameters were investigated. On average the DPOAE level increases with increasing stimulus levels, L_1 and L_2, up to near 65 dB SPL were it saturates. It has been found than decreasing L_2 below L_1 about 10 dB does not reduce the DPOAE level but increases its sensitivity to cochlear insult such as exposure to high levels of noise Sutton et al. (1994). Therefore, a maintaining a difference between the stimulus levels makes the DPOAE measure more clinically relevant. The separation between the stimulus frequencies, f_1 and f_2 ($f_1 < f_2$), also has an impact and is typically set as a ratio f_2/f_1. The DPOAE level exhibits a bell-shaped dependence on the stimulus ratio with a single maximum near 1.22 and a 3-dB bandwidth of about $0.1f_1/f_2$, although this bandwidth is typically not reported. The ratio that, for fixed stimulus levels, evokes the largest DPOAE level is referred to as the “optimal” ratio, while other senses of “optimal” could be imagined. There is broad consensus in the literature that the optimal ratio is 1.22 across frequency, but also that it is a slightly increasing function of increasing frequency (Harris et al., 1989). One study by Nielsen et al. (1993) indicated differently, that for stimulus frequencies below 1500 Hz the optimal ratio *decreases* as a function of decreasing frequency. A number of optimal-ratio studies also extend to stimulus frequencies below 1500 and they show the opposite.

The broad, smoothly varying average dependencies of the DPOAE level on the stimulus parameters do not always resemble dependencies in individual people. This is especially the case when the stimulus levels are lowered to be more clinically relevant, as argued above. It is well-known that the DPOAE level exhibits a rippling fine structure with 3-4 notches per octave of highly variable depths and soft maxima inbetween (Reuter and Hammershøi, 2006). The notches increase the variability around the average across individuals if they are not somehow taken into account in the analysis. It was recognized early on that the fine structure resembles the comb pattern of an interference between two sources that contribute simultaneously to the recorded DPOAE level. Several signal processing and selective suppression strategies exist to separate the contributions from two such sources but they rely on assumptions that may not always be justified.

The clinical relevance of DPOAE has also been studied extensively. The present review focused specifically on the ability of DPOAE to indicate elevated audiometric thresholds of minimum audibility. DPOAE levels, and the minimum stimulus level that evokes a DPOAE above the noise floor, identifies clinically normal and impaired listeners at relatively low rates of false alarms and misses. Broadly speaking, the relation is not unambiguous, because the DPOAE and audiometric thresholds do not rely exclusively on the same physiological structures and mechanisms. At stimulus frequencies below 707 Hz, and in some studies below 1000 Hz, the prediction of audiometric hearing status does not appear to be better than chance prediction because of significant difficulties in equating the noise levels across frequency. Noisy or lacking low-frequency data thus render the comparison of previous results below 1000 Hz to results above 1000 Hz unfair, or at least inconclusive (Stover et al., 1996a).

In the ear canal where the OAE is recorded there is noise at low frequencies. It may be steady noise from breathing and heartbeating, and noise of more sudden character, like swallowing, coughing, or movement that inhibits the measurement, because the OAE is naturally low in level, typically up to just 20 dB SPL in human adults. The noise can spread to higher frequencies when it spans a broad frequency band, or when it is not stationary, or both, within the time windows of analysis. Therefore, from the first report of OAE (Kemp, 1978), it has been customary to highpass filter the microphone output, if not by analog second- or higher-order highpass filters, then by using microphones that are only vaguely sensitive to low frequencies. The 3-dB cutoff frequency is typically between 250 and 500 Hz, and in some studies it varies with the frequency being measured (Gorga et al., 1994; Stover et al., 1996a). Highpass filtering is a simple and effective solution to a difficult noise problem that must be solved, if OAE measurements above the highpass cutoff frequency should not take too long (minutes per point) and render them impractical clinically.

The review had focus on studies that used the classical way of estimating the response and the noise by taking the FFT (fast Fourier transform) of consequtive time windows (Whitehead et al., 1994). The FFT decomposes the time signal into all the tones that fit exactly with an integer number of cycles into the window. Any signal component which is not periodic within the window cannot be represented by that finite number of tones. The limitation determines what stimulus and response frequencies can be used in a DPOAE measurement. The DPOAE response is estimated as the power in the one FFT bin at the $2f_1 - f_2$ frequency. There are at least two different ways of calculating the noise associated with the given $2f_1 - f_2$ frequency. In one, the summed power of one or more neighboring FFT bins is taken as a good-enough indication of the noise at the $2f_1 - f_2$ frequency. The method assumes that the noise is random across frequency and that, therefore, the powers at the $2f_1 - f_2$ bin and neighboring bins are independent but the same when averaged over the duration of the measurement. In the other way of calculating the noise, the DPOAE stimulus and response tones are cancelled out by subtracting temporally adjacent measurements. If stimulus and response were exactly the same in the two subtracted measurements, an estimate of how much the response changed due to noise between the two measurements can be obtained from the power *at* the $2f_1 - f_2$ frequency. If the noise in the DPOAE measurement is random and flat across frequency, and if the bandwidths of the estimates are equalized, then the two methods should be equivalent. If the measurements are, on the other hand, subjected to noise that is not stationary, such as that at low frequencies, their assumptions are violated and their results probably different. Since the existence of a DPOAE at a given frequency is always determined from a criterion threshold on the signal-to-noise ratio, it must be of importance for low-frequency measurements in particular to revisit noise calcuation, the rejection criterion, or both. These may, together with highpass filtering, have limited the measurement of low-frequency emissions.

Perhaps as a result of the reported difficulties in making low-frequency DPOAE measurements, published $2f_1 - f_2$ DPOAE data from humans extend down to an f_2 or geometric-mean frequency of 500 Hz. This low-frequency DPOAE has been measured for a wide range of stimulus levels and nearly always with the stimulus ratio fixed near 1.22. The ratio at 1.22 sets the $2f_1 - f_2$ frequency near to 333 Hz. It can be stated fairly confidently that DPOAE has not been studied systematically in humans for f_2 frequencies below 500 Hz or at $2f_1 - f_2$ frequencies below 300 Hz. DPOAE measurement for screening and clinical purposes in a wide frequency band uses standard stimulus parameters. The stimulus ratio is fixed near 1.2 and the stimulus level is 65/55 dB SPL. It is normal for the f_2 frequency to extend down to 750 Hz (Poling et al., 2014), but there is usually not a noteworthy mention of the low-frequency end of the spectrum. The literature reflects little attention to DPOAE for f_2 frequencies below 1000 Hz.

Some differences *have* been shown between DPOAE evoked by f_2 frequencies at 500-1000 Hz and DPOAE evoked by f_2 frequencies above 1000 Hz. Gorga et al. (2008) for instance presented 40-h long measurements of DPOAE-suppression tuning curves at f_2 frequencies 500 and 4000 Hz from normal-hearing humans. The DPOAE level exhibits a markedly broader tuning at an f_2 frequency of 500 Hz compared to that at 4000 Hz, and the low-frequency DPOAE level also has a smaller dynamic range for the same range of stimulus levels. These tuning characteristics arguably give the full picture of the ratio and level dependencies that have been observed over many years of study. Another difference between low- and higher-frequency DPOAE is seen in its phase, which is often not reported with the level. Level measures are more traditional in audiology and arguably also more intuitive to understand than phase. The phase normally rotates as a function of frequency below about 2000 Hz (Dhar et al., 2011; Shera et al., 2000). Above, it is approximately invariant, and this invariance is an assumption that underlies some signal processing methods to separate contributions from the two putative sources of DPOAE (basal source ignored) (Kalluri and Shera, 2001).

Fully utilizing DPOAE in assessment of cochlear activity requires a deeper understanding of the generating mechanisms. Low-frequency stimuli extend further toward the apex of the cochlea, and for these in particular, the literature suggests relatively large contributions from DPOAE sources that are basal to the peaks of the stimulus waves (Martin et al., 2003). Their contributions have been shown, for instance, to mask the ability of DPOAE to detect hearing loss in the regions of the basilar membrane where the stimulus waves peak. Basal sources, more clearly evident in low-frequency DPOAE, are currently not well incoorporated into the consensual two-source theory of DPOAE generation.

4 Goal of the present study

In summary, the review above elucidates a challenge and a limitation of reported DPOAE studies. These translate into goals of the present study and points of discussion later on.

(1) Handling low-frequency noise in the ear canal

Noise in the ear canal increases steeply with decreasing frequency below 300-500 Hz. The noise cannot be assumed to be stationary or similar in consequtive measurements. Highpass filtering with a cutoff frequency between 300 and 500 Hz can exclude the noise for measurements at higher frequencies. Alternatively, it must be coped with in another way than filtering for measurements *in* the low-frequency range.

(2) Measuring DPOAE *in* the low-frequency noise

No $2f_1 - f_2$ DPOAE data from humans are published for f_2 or geometric-mean frequencies below 500 Hz. Measurements that extend to 500 Hz tend to have the stimulus ratio f_2/f_1 fixed near 1.2 which sets the $2f_1 - f_2$ frequency to 333 Hz. While there is no indication that the DPOAE does not extend to lower frequencies, their existence has not been demonstrated for $2f_1 - f_2$ frequencies below 300 Hz.

5 Overview of experimental work

Four studies were carried out and documented in papers within the scope of the present study. Paper A reconsiders basic choices within the standard FFT-bin signal processing scheme to allow for measurements at specific f_2/f_1 ratios at low frequencies. Paper B describes the design of electroacoustic instrumentation to enable measurement of emissions at subclinical frequencies. Paper C relates $2f_1 - f_2$ distortion emission data around 1231 Hz to data around 246 Hz and, subsequently, Paper D presents data from 88, 176, and 264 Hz. The data show that the largest emission level is evoked at low frequencies when the f_2/f_1 ratio is markedly higher than at mid and higher frequencies. A relation between the optimal f_2/f_1 ratio and the ERB (equivalent rectangular bandwidth) is proposed.

More than half of the recorded data in study C and D has not yet been reported in papers. The unreported recordings from study D in particular are unprecedented needing more pondering before a proper presentation can be given. A summary of each paper is given below, each ending with a perspective toward the two overall goals of the present study. The full papers are included in the appendix.

Summary of Paper A: Avoiding spectral leakage in measurements of distortion-product otoacoustic emissions

Introduction

The FFT is normally used to extract the level and phase of stimulus and response tones from a DPOAE measurement. A finite number of tones fits with an integer number of cycles within each recording. The FFT has these tones available to represent the recorded time signal in the frequency domain. Recordings containing tones with frequencies completing a non-integer number of cycles cannot be represented exactly in the frequency domain and show instead as decaying skirts of artifact energy around the nearest frequencies. The misrepresented spectral component is then said to leak into the neighboring bins. In DPOAE measurements the stimulus frequencies must be selected to exactly complete an integer number of cycles within the duration of the recording to be transformed by the FFT.

Methods

In a DPOAE measurement the recording is made at a sampling frequency f_s and analyzed with the FFT in time windows of length N samples. The spectral components available in the FFT are then $k \cdot f_s/N$ where k is an indexing integer. To make sure that the frequencies of the stimulus and response tones in the DPOAE measurement always fall at spectral components available in the FFT, the stimulus frequencies can be specified by the index as k_1 and k_2 ($k_1 < k_2$). The response frequencies fulfill the integer requirement when the stimulus frequencies do. DPOAE measurements are normally made with reference to the decimal-numbered stimulus ratio f_2/f_1, where f_1 and f_2 are the stimulus frequencies. By specifying the ratio as the fraction k_2/k_1 instead, all stimulus and response frequencies can be derived without rounding or truncation from a given reference k such as $2k_1 - k_2$.

Results

In an example the typical ratio of 1.22 can be specified as the fraction $122/100 = 61/50$. The associated $2k_1 - k_2$ bin then has index $2 \cdot 50 - 61 = 37$, implying that DPOAE measurements can only be made exactly at a ratio of 1.22 without errors due to spectral leakage at every 37 bin in the FFT. A markedly higher resolution of 7 can be attained with the ratio 11/9 (1.222...), or of 4 and 3 with ratios 6/5 (1.20) and 5/4 (1.25). Only for a ratio of 3/2 (1.50) is the DPOAE resolution 1, that is, equal to the FFT resolution.

Fig. 1 shows simulated DPOAE measurements and the error on the DPOAE estimate due to spectral leakage from stimulus tones that are not selected according to the integer requirement. The error increases with decreasing frequency to more than 10 dB in the normal frequency range of measurement ($N = 4096$, $f_2/f_1 = 1.22$ and $L_1/L_2/L_{dp} = 65/55/0$ dB SPL). Lower FFT resolutions (smaller Ns) and lower-frequency

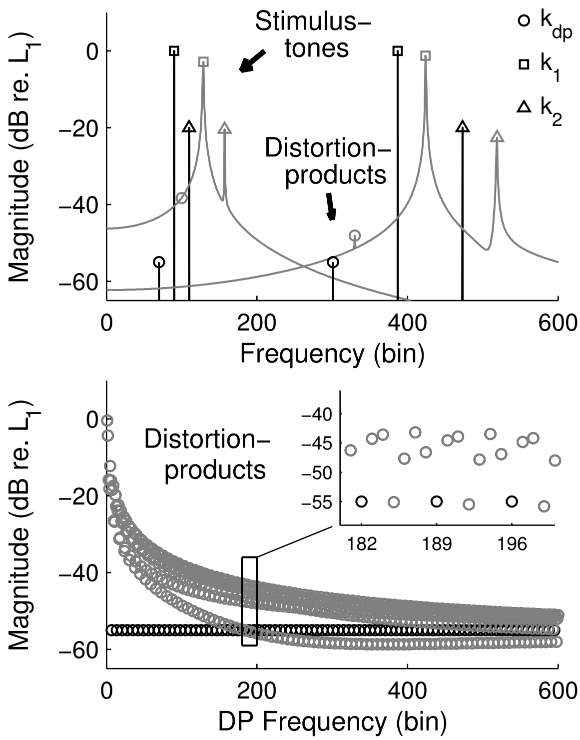

Fig. 1: Example result of simulated DPOAE measurement in study A. The top panel shows the FFT of two pairs of stimulus tones and their distortion tones. Skirts of spectral leakage occur in the gray curves because the stimulus frequencies fall between frequency bins in the FFT. For the black curves the stimulus tones have been selected according to the integer requirement. The bottom panel shows that the error due to spectral leakage increases toward low frequencies in fixed-ratio measurements. A similar trend toward low frequencies (not shown) can be seen in the error on the stimulus ratio when the stimulus frequencies are rounded to fall at frequency bins in the FFT.

measurements exhibit larger errors due to spectral leakage because the stimulus and response frequencies are closer.

Discussion

It stands to reason that spectral leakage must be avoided in DPOAE measurements. It is particularly challenging to do so when the stimulus ratio must be fixed across frequency, because the bins in the FFT are linearly spaced. If the stimulus frequencies are simply stepped the same number of bins, the stimulus ratio changes as the frequency is swept. Rounding or truncating the frequencies involved off to the nearest bins in the FFT solves the problem by allowing small variation in the stimulus ratio. Toward low frequencies however the error on the ratio due to this rounding increases systematically. The study here shows exactly how the stimulus ratio and the FFT resolution together dictate the highest DPOAE resolution at which measurements can be made without rounding errors on the ratio and spectral leakage.

Relation to goals of present study

Rounding errors on the stimulus parameters or spectral leakage are both, in a sense, noise induced by the FFT signal processing method. Accepting rounding errors on the stimulus parameters leaves poor control over the DPOAE experiment. The errors are not large, and probably insignificant, at the frequencies that are typically tested, but in low-frequency measurements using the FFT analysis, the errors can be significant. With the goal of measuring at low frequencies, neither rounding error nor spectral leakage could be ignored, and the study suggests a strategy that works around both. It also gives incentive to use higher FFT resolutions, as the potential errors are larger for low FFT resolutions.

Summary of Paper B: Design of an Acoustic Probe to Measure Otoacoustic Emissions Below 0.5 kHz

Introduction

In preparation for a forthcoming DPOAE study the transducer sensitivities and distortion of the ER10C probe system from Etymotic Research Inc. were measured. It has equalization filters that flattens the input sensitivity within about 5 dB from 100 to 10000 Hz. The microphone sensitivity alone (as inferred by measurements of an unequalized ER10B+ microphone) has a soft resonance around 2500 Hz and rolls off outside the frequency range from about 300 to 5000 Hz. The noise floor of the unequalized ER10B+ microphone increases steeply below 80 Hz. The loudspeakers in the ER10C probe system has third-harmonic distortion up to 40 dB below stimulus levels of 65 dB SPL, and such high distortion could potentially interfere with DPOAE measurement. These observations of the equipment that was available motivated the development of a probe system designed specifically for measurements at low frequencies.

Methods

The design targets for the input and output sensitivities, respectively, were flat from 30 to 2000 Hz and flat with as little distortion as possible up to 3000 Hz. Excessive noise was expected to set the low limit and the dimensions of the probe would cause resonances that are not easy to equalize at higher frequencies. The developed probe has dynamic loudspeakers with relatively large diaphragms of diameter 12 mm. More air can then be moved with less harmonic distortion. It has two parallel-connected pressure-gradient electret microphones with a flat sensitivity toward low frequencies. Electronic highpass filtering of the microphone output is necessary, but the cutoff frequency is much lower than normally seen, around 30 Hz. The transducers are assembled in a cut block of polyoxymethylene and coupled into the ear canal via narrow tubes of stainless steel.The probe is shown in two subjects in Fig. 2.

Results

The microphone in the custom-built probe has a sensitivity that is flat within a few dB from 10 Hz (not measured further down) to 1000 Hz. At higher frequencies it rolls off approximately at a rate of 40 dB/decade. Around 2500 and 9500 Hz it has two sharp resonances because of its placement near a rigid boundary. The resonances are unfortunate but not critical to the frequency range of interest. The probe microphone is markedly more sensitive than the unequalized ER10B+ probe up to about 300 Hz. From 300 to 3000 Hz it is slightly more sensitive and less sensitive at higher frequencies, except at the resonances.

The loudspeaker sensitivities resemble the frequency response of a Helmholtz resonator, as expected from the placement of the loudspeak-

Fig. 2: Custom-built acoustic probe in two subjects for measurement of low-frequency DPOAE. The electroacoustic design is detailed in Paper B and, to some extent, in Paper D. The design is somewhat unfinished but has worked and held its calibration within 1 dB throughout the use in 21 subjects in study D.

ers within small volumes that open into the ear via comparatively small holes. The soft resonances of the cavities are near 300-400 Hz. The custom-built probe loudspeakers have more than 20 dB less harmonic distortion compared to the ER10C loudspeakers when driven with voltages giving 65 dB SPL in a B&K 4157 ear simulator. However the probe microphone has relatively more $f_2 - f_1$ distortion for a reason that has yet to be identified.

Discussion

The custom-built probe is favorable for measurements below 1000 Hz. Its stimulus to the ear contains markedly less harmonic distortion and the microphone sensitivity is flat to very low frequencies. The benefits of the probe over the ER10B+ and ER10C systems are, however, only clearly apparent for measurements below about 300 Hz. In further work the resonance within the microphone tube should be reduced. This is likely the main reason for the excessive $f_2 - f_1$ distortion in measurements above 300 Hz.

Relation to goals of present study

Commercially available OAE measurement systems are not designed for low-frequency measurements. They are instead sensibly tailored for shorter and artifact-free measurements at higher frequencies. The custom-built probe reported here was developed mainly as a curiousity to see if it was realistic to condition low-frequency measurements with a higher signal-to-noise ratio than off-the-shelf probe systems do. Such recordings of higher quality can enable a more effective identification of "noisy" measurements, and thus support both goals of the present study.

Summary of Paper C: Stimulus ratio dependence of low-frequency distortion-product otoacoustic emissions in humans

Introduction

As a first step toward extending the measurement of DPOAE in humans to f_2 frequencies below 500 Hz, or $2f_1 - f_2$ frequencies below 300 Hz, the dependence of the DPOAE level on the stimulus ratio, and some other stimulus parameters, was measured around a well-studied mid frequency and a low frequency. This was done without major deviations from standard instrumentation and methodology.

Methods

Eighteen out of 21 recruited human subjects had normal hearing thresholds to be included in the study. Eight or nine stimulus ratios f_2/f_1 were measured while fixing f_2 at 513-545 Hz and at 2051-2180 Hz and $2f_1 - f_2$ at 246 and 1231 Hz. These are shown in Fig. 3. In the fixed-f_2 conditions the $2f_1 - f_2$ frequency varied around the frequencies where they were fixed in the fixed-$2f_1 - f_2$ conditions, and vice versa. The four conditions were measured at three different L_1/L_2 combinations 65/45, 65/55 and 70/60 dB SPL. The averaging duration was 9.6 and 4.1 s around 246 and 1231 Hz, respectively, with no automatic or manual noise rejection during the measurement. The noise was calculated in three different ways to indicate potential differences.

The study used the ER10C probe system from Etymotic Research Inc., because the low-frequency probe had not yet been considered. The ER10C microphone does not limit the measurement around 246 Hz, but the ER10C loudspeakers have harmonic distortion up to 40 dB below

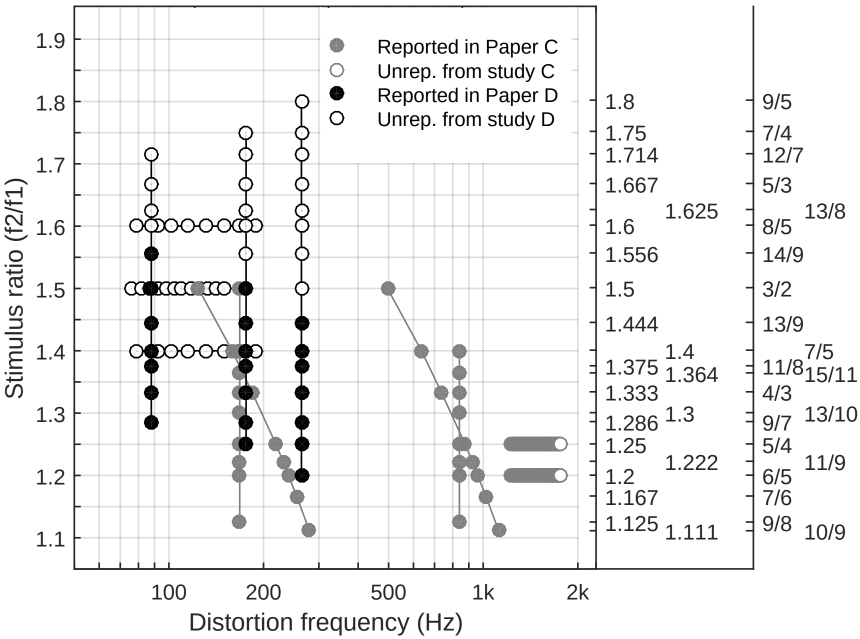

Fig. 3: Stimulus-frequency ratios at which measurements were made in the studies described in Paper C and D. The vertical axes to the right show the exact ratios as decimal numbers and fractions. The ratios were selected as described in Paper A. For all ratios above 1.5, the $f_2 - f_1$ distortion frequency was held fixed instead of the $2f_1 - f_2$ frequency. In the two angled lines of ratios, the f_2 frequency was fixed at 513-545 Hz and at 2051-2180 Hz. In the two horizontal lines of ratios between 1000 and 2000 Hz, the stimulus ratio was fixed.

the stimulus, as shown in Paper B. This harmonic distortion stimulates the ear basal to the f_2 place on the basilar membrane. The distortion is mainly third-harmonic and combination frequencies, between the third harmonic and the stimulus frequencies, do not fall at the $2f_1 - f_2$ frequency of interest. A possible indirect influence of the harmonic stimulus has not been studied systematically. The closest type of study is DPOAE measurement with high-frequency third stimulus tones intended to interfere with sources that are basal to the f_2 place (Martin et al., 2003).

Results

In both frequency ranges tested the DPOAE level is typically and on average a bell-shaped function of the stimulus ratio, shown in Fig. 4. Around 1231 Hz the bell peaks at a ratio of 1.238 whereas around 246 Hz it peaks at 1.333. At the optimal ratio the DPOAE level at the two frequencies is not different. The 246-Hz optimal ratio of 1.333 is larger than other optimal ratios found in the literature and it appears to arise as a result of the low measurement frequency. The optimal ratios do not change significantly at the three stimulus levels tested. It also does not make a significant difference whether f_2 or $2f_1-f_2$ are fixed, provided that they are swept within the same frequency range in the two paradigms.

The noise level is different in the two frequency ranges, typically 10 dB higher at the low than at the high frequency. In the fixed-$2f_1 - f_2$ measurements the noise is a flat function of the ratio, because the noise is estimated from the same FFT bin regardless of the ratio. In the fixed-f_2 measurements the noise is an increasing function of increasing ratio because the $2f_1 - f_2$ frequency decreases with increasing ratio when f_2 is fixed. Regarding the three ways of estimating the noise, the noise estimated from bins neighboring the $2f_1 - f_2$ frequency vary much less than the noise estimated *at* the $2f_1-f_2$ frequency in the difference signal between two temporally adjacent measurements.

Discussion

Considering that middle ear transmits less energy at low frequencies to and from the cochlea, it is a curious result that the same maximum DPOAE level can be evoked in the two frequency ranges. Either the middle ear has no average effect around 246 Hz, or the evoked energy is more at the low frequency by an amount that compensates for reduced middle-ear transmission.

The noise is different at the two frequencies because a rather naïve, but attractively simple, approach was taken with respect to noise in the measurements. No noise rejection was applied during the measurement and the averaging duration was doubled at the low frequency, not adjusted according to the measured noise floor as several other studies at-

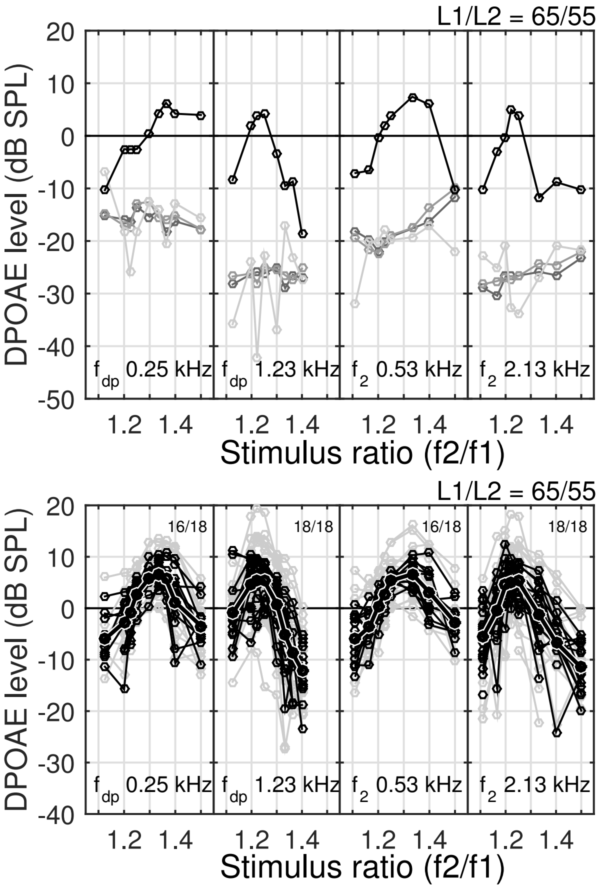

Fig. 4: Example results from study C. The top panels are from subject 10 and the bottom panels show the averages across subjects in the stimulus condition with stimulus levels at 65/55 dB SPL.

tempt. This strategy provides uninterrupted measurements of the noise and the response at the risk of obtaining only few measurements without excessive noise. The outcome, however, was that nearly all tested subjects with normal hearing had DPOAE levels 3 dB above the noise level, most often in both frequency ranges.

The larger variability of the noise calculated from the difference signal was initially thought only to be a result of the smaller bandwidth. It may also be related to a later observation that it makes a difference on the estimated noise exactly which repeated measurements are subtracted from each other. The noise estimate is volatile perhaps because some underlying assumptions on the character of the noise are not fulfilled (see page 24). In any case it seems impractical and may have caused some of the problems encountered by e.g. Gorga et al. (1994); Stover et al. (1996a) in detecting low-frequency DPOAE according to a criterion on the signal-to-noise ratio.

The optimal-ratio data were compared to data from five previous studies (eight in study D). It was argued that the optimal ratio may be proportional to a measure of auditory frequency tuning such as the ERB, derived from psychophysical tuning curves (Glasberg and Moore, 1990). All data from a total of 79 subjects appear to be well-guided by 1.516 ERB around the f_2 frequency. The only data in the low end of the spectrum is from the present study, and this is where the optimal ratio changes the most as a function of frequency, if it is in fact proportional to the ERB. Study D solidifies the relation further.

Relation to goals of present study

No particular effort was made to handle low-frequency noise, except the subjects were asked to sit quietly, and try not to swallow, and the averaging duration was increased from 4.1 s around 1231 Hz to 9.6 s around 246 Hz. This strategy was intended to give a realistic impression of the problems described in the literature regarding difficulties in making low-frequency measurements. A strategy that attempted to equalize the noise level in both frequency ranges tested may or may not have been very time consuming in the subjects, but measurements of the level-ratio curves do not appear to require such equalized noise floors. The study indicated that noise estimation from the difference signal may not be feasible because of high variability, perhaps induced by nonstationary noise sources. The study also provided a first indication that it is necessary, rather than optional, to increase the stimulus ratio toward low frequencies to maintain the maximum DPOAE level and thus enable DPOAE measurement above the noise floor.

Summary of Paper D: Distortion-product otoacoustic emission measured below 300 Hz in normal-hearing human subjects

Introduction

Paper C compared the DPOAE level-ratio dependence at a well-known mid frequency, near 1231 Hz, to that at a largely unknown low frequency, near 246 Hz. Paper D extends the systematic study of the DPOAE level-ratio dependence to still lower frequencies, 87.9, 176 and 264 Hz, and it also reports the DPOAE phase-ratio dependence. To make it possible some methodology deviates from the standard.

Methods

Twenty-one out of 34 recruited human subjects were included in the study (one subject was also in study C). The exact ratios measured for $2f_1 - f_2$ fixed at 87.9, 176 and 264 Hz are shown in Fig. 3. The stimulus levels were 65/55 dB SPL. The low-frequency probe reported in Paper B was used in this experiment to condition the low-frequency measurements similarly to those at higher frequencies.

Pilot study for this experiment determined that, within equisized bands of the FFT, the noise level in the ear canal increased more than 40 dB from 300 to 30 Hz. As a result, the time spent measuring at a single frequency increased a factor 4 per octave. The resulting averaging duration was 95.7, 24.7 and 9.6 s, respectively.

Subjects were given time to relax and asked to sit as quietly as possible. They were allowed to read on a tablet device and given breaks every 10-15 min to take deep breaths, clear the throat, and so on. Single measurements were rejected, if they contained noise of sudden character, or aperiodicity, for instance as a result slowly varying noise.

Instead of estimating the noise from the difference signal between consequtive measurements of from equisized sidebands around the distortion frequency, the noise was calculated from the outer two thirds of an ERB around the distortion frequency. This gives a weighting of the sideband energy that relates to hearing and, to some extent, the increasing noise floor toward low frequencies.

Results

The DPOAE dependence on the stimulus ratio at 264 Hz could be measured in 20 out of the 21 subjects included in the experiment. The prevalence was determined by a 3-dB criterion on the signal-to-noise ratio and it decreases toward lower frequencies because the noise level increases, not because the average DPOAE level decreases. At 176 Hz the dependence could be measured in 15 subjects and at 87.9 Hz it could be measured in 8 subjects.

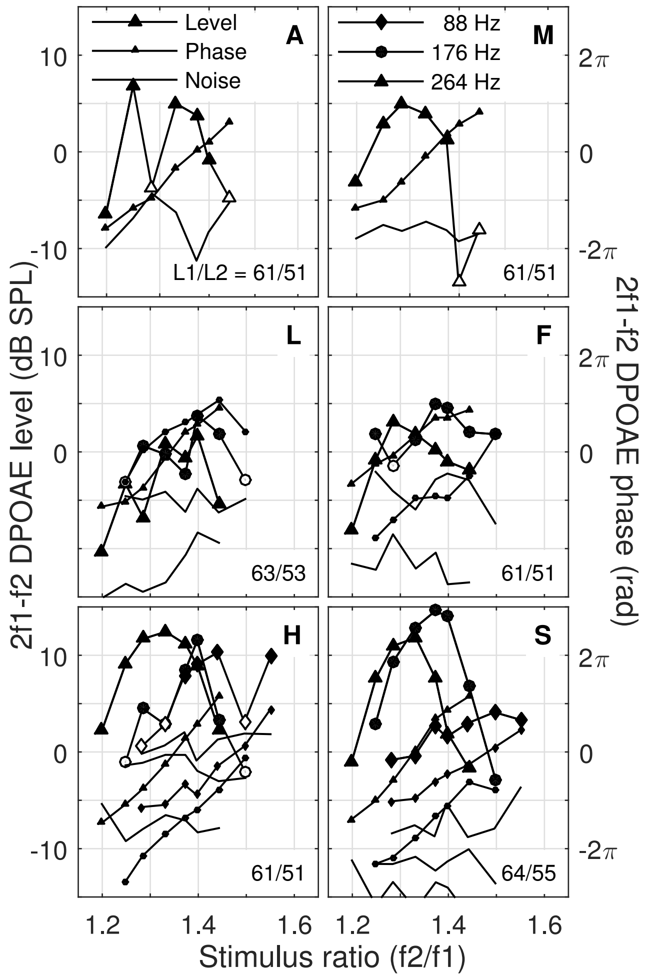

Fig. 5: Example DPOAE results from six individuals in study D.

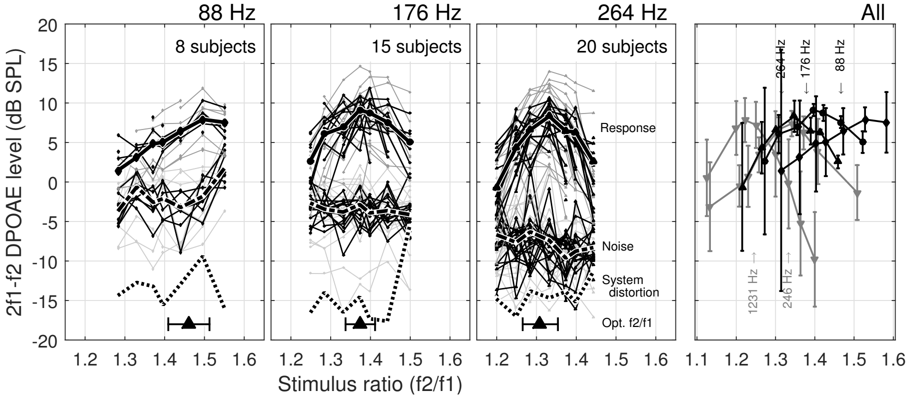

Fig. 6: Average DPOAE-level results in study D. The gray curves in the rightmost subfigure are for comparison the average measurements in study C. Triangles are placed at the optimal ratio calculated as the average of all individual optimal ratios.

The DPOAE level-ratio dependence is typically and on average bell shaped with a single maximum for the $2f_1 - f_2$ frequencies 176 and 264 Hz, shown in Fig. 5 and 6. The bells are progressively broader toward low frequencies. At 87.9 Hz only one side of the average bell is convincingly present within the range of ratios measured, and it is not certain that the lowest-frequency DPOAE level-ratio dependence is in fact bell shaped. The dependencies peak at optimal ratios 1.461, 1.376 and 1.331 for 87.9, 176 and 264 Hz, respectively. The peak levels are approximately equal at all three frequencies, and also to the peak levels expected at higher frequencies.

The DPOAE phase is an increasing function of increasing stimulus ratio. Within the bell of the level-ratio dependence, the phase increases approximately one cycle of the $2f_1 - f_2$ frequency and the slope is shallower toward lower frequencies. At the optimal ratio the phase crosses approximately 0, π and 0 radians at the frequencies, respectively.

Discussion

In line with the results of study C, the bell-shaped DPOAE level-ratio dependence broadens and the optimal ratio increases with decreasing frequency. The result in Fig. 7 strongly supports the suggestion in study C

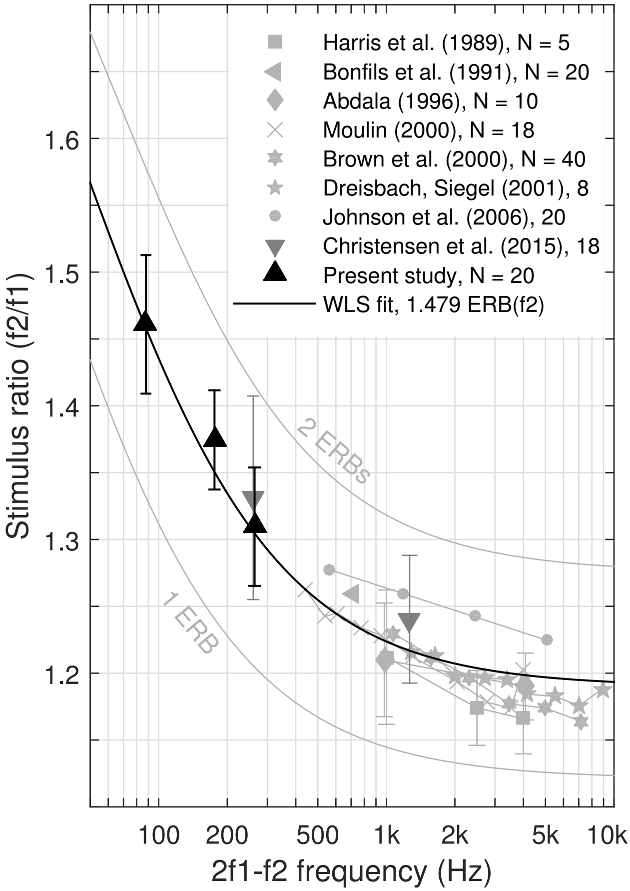

Fig. 7: Dependence of the stimulus ratio evoking the largest DPOAE level as a function of the $2f_1 - f_2$ distortion frequency. The optimal-ratio data are plotted with data from eight previous studies as reviewed in paper C and D.

that the optimal ratio is proportional to the ERB which increases linearly with increasing frequency (Glasberg and Moore, 1990). By inclusion of the new data the proportionality factor changes from 1.516 to 1.479.

On average and in several individual subjects, the maximum DPOAE level was highest at 176 Hz and lowest at 87.9 Hz, but in general the maximum levels seen were similar at all three frequencies. As also discussed in study C, this unattenuated level toward low frequencies is not consistent with the view that the DPOAE rolls off as the middle ear is less able to transmit low frequencies to and from the cochlea. It may be that a compensating amount of distortion energy is generated in the apical turns of the cochlea. An increased spread of stimulus excitation toward the base could be integrating more basal sources of DPOAE.

The DPOAE phase increases with increasing stimulus ratio, perhaps because the places of maximum vibration, due to the stimulus tones, move closer toward the base. The phase crossing 0, π and 0 radians at the optimal ratio resembles the phase of a second-order resonance, when two sources contribute to the same measurement. This two-source hypothesis is further supported in the subjects who display the largest DPOAE level at 176 Hz where the phase crosses π radians. It is possible that the basal-to-f_2 source has a phase that rotates relative to the phase of the source related to the f_2 place of maximum vibration.

At low frequencies a significant basal contribution seems more plausible than a contribution from the reflection at the $2f_1 - f_2$ place. The $2f_1 - f_2$ wave is not expected to peak sharply in the apex and this is necessary to produce a significant reflection (Zweig and Shera, 1995). Another DPOAE component at the $f_2 - f_1$ frequency may also contribute toward low frequencies when the measurement uses a stimulus ratio that changes proportionally to the ERB. Below 100 Hz in particular the optimal ratio crosses 1.5, and at this ratio, the $2f_1 - f_2$ frequency is equal to the $f_2 - f_1$ frequency. To the extent that both components are present at ratios in the neighborhood of 1.5, interaction between the two at 1.5 must be expected. This experiment included measurements to study interactions between the two distinct DPOAE components, but more pondering is necessary before a proper data presentation can be given.

Relation to goals of present study

Several efforts to handle low-frequency noise in the ear canal allowed for relatively quick measurement of DPOAE at lower frequencies than has been reported previously. These efforts included subject instruction, increased averaging duration according to the average increase of the noise floor (determined in pilot experimentation), and rejection of measurements that were noisy in a certain sense to ensure a specified number measurements without them. To an extent at the lowest frequencies, the custom-built probe system also made the measurements possible. The necessary efforts were identified over an extended period of time of experimentation in single subjects, after study C had been carried out.

The noise levels did, however, not end up equalized at the three frequencies tested. They were low enough to show convincingly that human DPOAE extends with the same maximum level to $2f_1 - f_2$ and f_2 frequencies 1-2 octaves below previous measurements. The bell-shaped dependence of the DPOAE level on the stimulus ratio is also maintained, but it broadens and the optimal ratio increases with decreasing frequency.

To measure DPOAE at a constant average level toward low $2f_1 - f_2$ frequencies, it is necessary to increase the stimulus ratio proportionally to the frequency dependence of the ERB. With support from the results of the studies in paper A, B and C, the study reported in this paper D fulfills both goals of the overall study, to handle low-frequency noise in the ear canal and measure DPOAE at low frequencies. Further details of the results in support of the goals are given in the following section.

6 Summary of results

The results of the present study are both methodological and phenomenological. Its goals presented in section 4 also reflect this distinction. The two sections below summarize the methodology applied to handle low-frequency noise in the ear canal and the phenomenology of the observations made of DPOAE in that low-frequency noise.

6.1 Handling low-frequency noise in the ear canal

Acoustic noise in the ear canal increases more than 40 dB in the low-frequency range from 300 to 30 Hz. Respiration and blood circulation, for instance, radiate low-frequency acoustic energy to the recording microphone in the enclosed ear-canal volume. The nonstationary nature of these noise sources poses a challenge in the methodology of OAE measurement. The microphone output cannot be highpass filtered, as in previous studies, because it also affects the low-frequency emission of interest. A reduced microphone sensitivity toward low frequencies is also detrimental to the signal-to-noise ratio but can be countered by extending the averaging duration. The low-frequency probe used in the present study has minimal highpass filtering above 30 Hz. A number of efforts enabled the recording of low-frequency emissions at all, even relatively quickly, despite the presence of nonstationary, low-frequency noise.

(1) Instruction of subjects

Subjects were seated in a reclining chair. The chair had to be soft, not only to make the subjects comfortable, also to absorb heartbeat energy bouncing off their backs and into the recording microphone. The subjects were given time to calm down and breath slowly. They were asked to breath mainly through the mouth as pilot experimentation indicated that air flow through the narrower nose was more harmful to the noise floor than air flow through the mouth. Subjects were also allowed to read on a tablet device and given short breaks approximately every 10-15 min to sip water and clear the throat.

(2) Size of analysis window and repetition rate

Recordings were repeated and analyzed continuously at a rate of 1.46 Hz. The repetition rate gave a high-enough frequency resolution at low frequencies. Beats of the heart were also unlikely to synchronize. In repeated measurements they are at different temporal places and averaging cancels them out.

(3) Rejection of "noisy" measurements

Very low-frequency noise (<30 Hz) occurs mainly as a result of breathing and minor movements of the probe with respect to the subject. Such noise shows itself as frequent and large jumps in the FFT, similar to the FFT of a unit step signal, and it is critically detrimental to the signal-to-noise ratio at low frequencies. A method was applied to identify such jumps, reject measurements containing them and extend the total measurement with the number of rejected measurements. The jumps could be identified as aperiodicity in the time domain from the difference between the first and the last sample in the analysis window. The crest factor, that is, the ratio between the peak and root-mean-square power, of the time signal was also not allowed to exceed 3. It is in this sense that measurements are referred to as “noisy” in the present study. Typically, the measurement was extended 10% due to this rejection strategy, but extreme cases of no and total rejection were also seen.

(4) Frequency-dependent averaging duration
To counter the overall increase in the noise floor, the number of repeated measurements was set to 8.19 s at 300 Hz and then increased a factor 4 per octave toward low frequencies. This was not enough to equalize the noise level at the three frequencies measured in study D. It is possible that the noise rejection strategy applied could have identified a different kind of noisy measurements to improve the average measurement. Beyond doubt, however, was the strategy applied better than none.

(5) Frequency-dependent noise estimation
In the analysis of the measurements the noise levels were calculated from the outer two thirds of an ERB around the distortion frequency. The noise level is normally calculated by summing the power in frequency bands that are said to represent the noise in the $2f_1 - f_2$ frequency band because they are near to it. But these bands are not changed as a function of the measured frequency. According to the ERB, the sense of what frequency bands are “near to” $2f_1 - f_2$ changes as a function of frequency and this was the main reasoning behind using it in calculating the noise level. Moving the noise bands closer to the $2f_1 - f_2$ frequency, while also narrowing them, accommodates very low-frequency measurements better, because the noise not flat across frequency. The noise is instead much higher to one side of $2f_1 - f_2$ than it is to the other side. The ERB-guided noise bands also accommodate well the ERB-based optimal-ratio relation suggested in Papers C and D.

(6) Stimulus delivery with little harmonic distortion
In addition to these strategies to improve the recorded signals, the low-frequency probe stimulated the ears with tones of higher quality than tones delivered by the ER10C probe system, which was also available. The harmonic distortion was lower than 0 dB SPL with the low-frequency probe and about 10-30 dB SPL with the ER10C, when stimulating with levels of 65 dB SPL. The low-frequency probe is, however, somewhat unfinished despite the fact that it holds its calibration well. For a reason that has yet to be identified, its microphones produce more distortion at the $f_2 - f_1$ frequency than the ER10C probe.

6.2 Distortion emission *in* low-frequency noise

At f_2 frequencies above about 600 Hz, the $2f_1 - f_2$ DPOAE level from human ears has previously been known to depend systematically on the ratio between the stimulus frequencies. The level is approximately bell shaped as a function of the ratio. As summarized in Fig. 7, the present study has, as its main result, extended the systematic study of the DPOAE dependence on the stimulus ratio to 87.9 Hz. The level-ratio dependence is also bell shaped at low frequencies. The ratio evoking the largest DPOAE level increases systematically with decreasing frequency. The trend is well approximated by 1.48 ERB around the f_2 frequency. Further perspectives of these results are discussed below.

(1) Low-frequency extent of DPOAE
The human ear emits $2f_1 - f_2$ distortion down to a frequency of at least 87.9 Hz. It does so at the same overall level as at higher frequencies. The distortion must be expected to go markedly lower in frequency. Considering that the middle ear is decreasingly able to transmit low frequencies (Marquardt et al., 2007; Puria, 2003), it is a rather curious result that the same distortion level from the cochlea can be evoked. It is possible that the reduced transmission is countered by the increased spread of excitation on the basilar membrane recruiting more cochlear sources of distortion basal to the f_2 place of maximum vibration (Har-

ris, 1990; Martin et al., 2003). This may be further investigated by comparing low-frequency emission measurements to a measure of the middle-ear transmission. One such measure using modulation of high-frequency DPOAE by lower-frequency tones was studied by Marquardt et al. (2007).

(2) Frequency-dependent optimal ratio

To obtain the same overall distortion level at low frequencies the ratio between the stimulus frequencies must vary as a linear function of frequency. This counters previous recommendations for clinical utility that the stimulus ratio be fixed at 1.22 across frequency (IEC:60645-6, 2009; Lonsbury-Martin and Martin, 2007). Specifically, combining previous with the present data, the ratio evoking the largest distortion level is proportional to an ERB, derived from phychophysical tuning curves and a linear function of the center frequency (Glasberg and Moore, 1990). The relation between the ERB and the optimal stimulus ratio is probably rooted in the frequency-dependent spread of excitation on the basilar membrane, where distortion is generated by active shaping of vibrations, as also discussed by Harris et al. (1989).

(3) Current practical low-frequency limit of DPOAE

The measurement of $2f_1 - f_2$ distortion from the ear is not practically impossible but methodologically demanding in the low-frequency range from 30 to 300 Hz. More so than in higher-frequency measurements. The measurement is increasingly time consuming toward low frequencies. The averaging duration must increase at least a factor 4 per octave to counter the increasing noise level. With today's methodology and understanding of DPOAE, a frequency of 100 Hz might constitute a soft low-frequency limit of practically, that is to say clinically, feasible measurements of DPOAE. The main reasons are, the microphone sensitivity of commercial systems rolls off significantly below 100 Hz, the methodology related to low-frequency noise rejection needs development beyond the present study, and, as described in the next point (4), $2f_1 - f_2$ DPOAE data are possibly influenced by $f_2 - f_1$ DPOAE below 100 Hz.

(4) Possible influence of $\mathbf{f}_2 - \mathbf{f}_1$ interference below 100 Hz

The present study has also identified $2f_1 - f_2$ distortion from the ear below 100 Hz to be complicated by the influence of simultaneous $f_2 - f_1$ distortion from the ear. The influence has yet to be understood and reported. However, when the measurement follows the optimal ratio, it crosses 1.5 around 74 Hz. At this ratio the $2f_1 - f_2$ frequency is equal to the $f_2 - f_1$ frequency and interference, modulation, or both, must be expected. Composite analysis of both level and phase of these two distinct distortion components is probably necessary to obtain a meaningful impression of the state of activity in the cochlea associated with low-frequency hearing.

7 Implications of results

For DPOAE measurement with f_2 frequencies above 1000 Hz a steep highpass filter at 250-500 Hz alleviates the problem of excessively long measurement durations due to low-frequency noise in the ear canal. For f_2 frequencies below 1000 Hz, however, that same highpass filter may inhibit the detection of noisy measurements and DPOAE by reducing the “noise”-to-noise and signal-to-noise ratio, respectively, below the cut-off frequency. The present study showed that it is possible to measure DPOAE down to at least 87.9 Hz with a highpass filter set at just 30 Hz, provided that an implemented noise-rejection strategy targets aperiodic, slowly varying and sudden burst noise in the recorded signal.

The methodology applied to measure down to 87.9 Hz here may be useful in obtaining data for f_2 frequencies between 500 and 1000 Hz where it has previously been difficult to obtain data. Studies attempting to relate DPOAE levels and thresholds to audiometric hearing status were reviewed specifically. These studies, despite long averaging durations, were not convincing for f_2 frequencies below 1000 Hz, and the authors suggested that their methodology was insufficient in their low-frequency measurements. The present study also experimented with the way these studies estimated the noise from the signal calculated as the difference between consequtive measurements (see page 24). The results showed that the noise estimate is probably too volatile for reliable rejection of single measurements based on their signal-to-noise ratio. The estimate exhibits much variation from measurement point to measurement point and, what is more, it depends on which measurements are subtracted from each other. These observations may, however, account only for the variation in noise estimates at relatively low frequencies where the noise is not stationary.

There are other phenomena related to low-frequency DPOAE than its relation to audiometric thresholds. With reference to the review in section 2, examples include basic dependencies on the stimulus parameters and tuning that change toward low frequencies, and more elusive characteristics that are still not well understood, such as the phase slope and contributions from sources that are basal to the f_2 place. A more complete picture of these characteristics may be obtained by extending the DPOAE measurement toward lower frequencies than is usually done.

In clinical settings it is of interest to sweep the state of the entire length of the cochlea with an objective measure. The DPOAE is one such measure, swept with the stimulus levels fixed at 65/55, the stimulus ratio fixed at 1.22 and the stimulus frequencies varying from 500-1000 Hz to 4000-8000 Hz. The present study indicates that the methodology may allow for clinically feasible measurements at frequencies lower than 500 Hz. It appears necessary, however, to change the stimulus ratio proportionally to an ERB around the reference frequency. The specific relations are reported in Paper D and they are, with f_2 [Hz] given,

$$\begin{aligned} f_1 &= f_2 - 1.479\mathrm{ERB}(f_2) && (1) \\ &= 0.8404 f_2 - 36.53 \quad \text{[Hz]} && (2) \end{aligned}$$

And with $2f_1 - f_2$ [Hz] given,

$$\begin{aligned} f_1 &= 2f_1 - f_2 + \frac{1.479}{1 - 2 \cdot 0.1079 \cdot 1.479}\mathrm{ERB}(2f_1 - f_2) && (3) \\ &= 1.2345 \cdot (2f_1 - f_2) + 53.66 \quad \text{[Hz]} && (4) \end{aligned}$$

To also fulfill the integer requirement from Paper A, the parameters can, when using FFT-based signal processing, be approximated with fractions and integers. Increasing the stimulus ratio toward low frequencies, to follow the one giving the largest average DPOAE level, has two potentially important implications that need further study. First, as the optimal ratio gets close to 1.5 between 50 and 100 Hz, the $2f_1 - f_2$ frequency is close to the $f_2 - f_1$ frequency and the distinct DPOAEs at these frequencies may interact. Second, when the $2f_1 - f_2$ frequency is swept and the optimal ratio is increased, the stimulus frequencies change progressively less relative to the change in the distortion frequency. This means that the f_2-related regions of DPOAE generation "stall" on the basilar membrane while the $2f_1 - f_2$ continues toward the apex. If instead f_2 is swept while the optimal ratio increases, the $2f_1 - f_2$ frequency moves progressively faster toward the apex than the stimulus frequencies.

8 Concluding remarks

The present study revised a basic parameter dependence of $2f_1 - f_2$ distortion emission from the human ear. It related previous physiological emission data to a well-known psychoacoustical model. The relation, and the emission phenomenon in general, extends to frequencies 1-2 octave lower than has previously been shown.

The study alludes to the application of low-frequency distortion emission to assess the active process in the apical cochlea, for the study of apical-basal differences, for cross-species comparisons, or for clinical assessment of low-frequency hearing. A specific working hypothesis is that the apical cochlea can actively suppress the sensation of low-frequency sound. Loss of such suppression mechanism could result in hypersensitivity to low-frequency sound and increased low-frequency masking or modulation of higher-frequency sound. Perhaps the present report will inspire a further exploration of these potentials.

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Paper A

Anders T. Christensen, Rodrigo Ordoñez and Dorte Hammershøi, “Avoiding spectral leakage in measurements of distortion-product otoacoustic emissions,” *Proceedings of Forum Acusticum 2014. European Acoustics Association*, September 2014.

Paper B

Anders T. Christensen, Rodrigo Ordoñez and Dorte Hammershøi, “Design of an Acoustic Probe to Measure Otoacoustic Emissions Below 0.5 kHz,” *Audio Engineering Society 58th International Conference: Music Induced Hearing Disorders*, June 2015.

Paper C

Anders T. Christensen, Rodrigo Ordoñez and Dorte Hammershøi, “Stimulus ratio dependence of low-frequency distortion-product otoacoustic emissions in humans,” *Journal of the Acoustical Society of America*, vol. 137, no. 2, pp. 679–689, 2015.

Paper c

Anders T. Christensen, Rodrigo Ordoñez and Dorte Hammershøi, “Erratum: Stimulus ratio dependence of low-frequency distortion-product otoacoustic emissions in humans [J. Acoust. Soc. Am. 137(2), 679 (2015)],” *Journal of the Acoustical Society of America*, accepted, in press, 2015.

Paper D

Anders T. Christensen, Rodrigo Ordoñez and Dorte Hammershøi, “Distortion-product otoacoustic emission measured below 300 Hz in normal-hearing human subjects,” *Journal of the Association for Research in Otolaryngology*, submitted, 2015.

Paper E

Anders T. Christensen, Rodrigo Ordoñez and Dorte Hammershøi, "Relating $2f_1 - f_2$ distortion product otoacoustic emission and equivalent rectangular bandwidth," *Proceedings of the International Symposium on Auditory and Audiological Research*, accepted, in press, August 2015.

Paper F

Anders T. Christensen, James Dewey, Sumitrajit Dhar, Rodrigo Ordoñez and Dorte Hammershøi, “A Pilot Study of Phase-Evoked Acoustic Responses From the Ears of Human Subjects,” *Proceedings of the 12th International Mechanics of Hearing Workshop*, June 2014.

SUMMARY

The sensory organ of hearing, the cochlea, emits faint sound as it processes incoming sound. Measurement of such ”otoacoustic emission” in the ear canal provides evidence for how the live, healthy ear works. Emissions at mid frequencies associated with speech is usually of prime interest. Low-frequency hearing has not yet been characterized by measurement of low-frequency emissions from the cochlea. Low-frequency emissions are expected to be covered in sounds of breathing, blood circulation, and so on, if they exist at all at measurable levels. The present study shows, in essence, that the human ear emits distortion at least 1-2 octaves lower in frequency than has previously been shown. The emission is promising for further exploratory and clinical assessment of cochlear activity associated with low-frequency hearing.

Anders received his M.Sc. degree in acoustics in 2012 from Aalborg University. From 2012 to 2015 he was employed at Aalborg University, pursuing a PhD degree in the field of otoacoustic emissions. Anders was a visiting research student at the National Center for Audiology, Western University, in Canada, the Fraunhofer Institute for Integrated Systems in Germany, and the Auditory Research Laboratory, Northwestern University, Evanston, IL, USA.

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