Impact of infrasound on the human cochlea

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Abstract

Low-frequency tones were reported to modulate the amplitude of distortion product otoacoustic emissions (DPOAEs) indicating periodic changes of the operating point of the cochlear amplifier. The present study investigates potential differences between infrasound and low-frequency sounds in their ability to modulate human DPOAEs. DPOAEs were recorded in 12 normally hearing subjects in the presence of a biasing tone with \( f_B = 6 \) Hz and a level \( L_B = 130 \) dB SPL. Primary frequencies were fixed at \( f_1 = 1.6 \) and \( f_2 = 2.0 \) kHz with fixed levels \( L_1 = 51 \) and \( L_2 = 30 \) dB SPL. A new measure, the modulation index (MI), was devised to characterise the degree of DPOAE modulation. In subsequent measurements with biasing tones of \( f_B = 12, 24 \) and \( 50 \) Hz, \( L_B \) was adjusted to maintain the MI as obtained individually at 6 Hz. Modulation patterns lagged with increasing \( f_B \). The necessary \( L_B \) decreased by 12 dB/octave with increasing \( f_B \) and ran almost parallel to the published infrasound detection threshold. No signs of an abrupt change in transmission into the cochlea were found between infra- and low-frequency sounds. The results show clearly that infrasound enters the inner ear, and can alter cochlear processing.

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1. Introduction

In the general understanding and in its definition infrasound stimuli are not audible for humans. The lowest audible frequency is often given in the region of 15–20 Hz. However infrasound is reported to be sensed by the ear (Jerger et al., 1966; Möller and Pedersen, 2004; Leventhall, 2007), and that the percept of sound within this frequency range changes gradually. Thus, in contrast to the upper frequency limit, the lower frequency limit of hearing is perceptionally and also psychophysically difficult to define, and a distinction between infrasound (IF) and low frequency (LF) sound seems hardly justifiable. One goal of the study is to see whether the cochlear acoustic emissions in this frequency range provide reason to define such a boundary.

Various publications report annoyance, deficiencies in mental performance, stress, vertigo or imbalance, and sleep disorders as consequence to overexposure to very-LF noise, in general. Furthermore, the vestibular system, skin, cardiovascular and respiratory systems, as well as inner
organs can be affected (Harris et al., 1976; Møller and Pedersen, 2004). Today, it appears to be clear that even down to a few Hz the inner ear is the most sensitive organ detecting LF sound (Møller and Pedersen, 2004). Effects on other organs seem to be presumably secondary and stress related. Their occurrence, however, makes it important to understand the pathways in which very-LF sound impacts on the body, and in this study, we focus on the sensation by the cochlea.

With the aim to observe the effect of very-low frequency sound on the cochlear partition, we analyse the modulation pattern of distortion product otoacoustic emissions (DPOAEs) which can be non-invasively recorded in the ear canal. Differences in cochlear acoustics might explain reported individual differences in susceptibility to infrasound.

Otoacoustic emissions (OAEs), as an epiphenomenon of hearing, are a by-product of the cochlear amplification process and hence can serve as a means of acquiring information about its state. DPOAEs arise directly from the compressive non-linearity of outer hair cells (OHCs) (Brownell et al., 1985; Kemp, 1986). The knowledge about the DPAOE generation process allows the indirect assessment of losses in sensitivity, frequency-selectivity, and compression in human ears (e.g., Kummer et al., 1998; Janssen et al., 2006).

Beside the evaluation of cochlear integrity, DPOAEs are a non-invasive means for assessing the motion of the cochlear partition. Frank and Kössl (1996, 1997) modulated DPOAEs by infrasound and low-frequency tones in gerbils and used the resulting DPOAE modulation pattern as an objective measure to evaluate the characteristics of the cochlear amplifier. Depending on the used frequency ($f_B = 25$ Hz) and sound pressure levels ($L_B = 95–125$ dB SPL) of the biasing tones, different DPOAE patterns occurred, and were interpreted as a result of periodic shifting of the operating point of this amplifier. Further studies used biasing techniques in animals to study the DPOAE generation process and the operation characteristics of the cochlear amplifier (Bian et al., 2002, 2004; Bian, 2004, 2006; Lukashkin and Russel, 2005).

In human subjects, Zwicker (1981) systematically analysed the suppression of click-evoked OAEs by a 62 Hz suppressor tone of 95 dB SPL and published so-called suppression period pattern. Scholz et al. (1999) recorded DPOAEs in normally hearing adults modulated by a biasing LF tone with a frequency of 32 Hz and a level of up to 115 dB SPL for various primary tone frequencies and levels. In a follow-up study, Hirschfelder et al. (2005) investigated a possible clinical application of modulated DPOAEs in a group of 20 patients suffering from endolymphatic hydrops. A 25 Hz biasing tone of 110 dB SPL (86 phone) modulated the DPOAEs of these patients to a significantly smaller degree compared to an age-matched control group of 20 normally hearing subjects. Marquardt et al. (2007) modulated DPOAEs by LF tones in the frequency range between 15 and 450 Hz in humans and guinea pigs. They adjusted the LF tone levels to maintain equal DPOAE modulation depth. The resulting so-called distortion product iso-modulation curves matched the human equal loudness contours (ISO 226: 2003) except for a down-up irregularity between 40 and 60 Hz. In guinea pigs this pronounced irregularity was located between 100 and 140 Hz.

The purpose of the present study was to investigate the impact of IF on the cochlear partition by using DPOAE modulation patterns as a non-invasive test for monitoring changes in cochlear sound processing, and a possible difference between the impact strength of IF and LF sounds.

2. Materials and methods

2.1. Subjects

Twelve volunteers (10 females and 2 males aged from 19 to 56 years, median = 25.5 years), who gave their informed consent, participated in this study. The procedure was approved by the ethics committee of Charité, Universitätsmedizin Berlin, in accordance with the declaration of Helsinki. All subjects had normal hearing, as verified by a pure-tone audiogram between 125 Hz and 8 kHz, their audiometric thresholds being within the range of 10 dB above or below the mean age-corrected thresholds (ISO 7029: 2000). Middle ear and retro-cochlear disorders were excluded by tympanometry and click-evoked auditory brainstem responses. None of the subjects had a history of otological or neurological disorders. No tinnitus was reported and no hyperacusis was found, as tested by the discomfort threshold which had to be above 75 dB HL (SchAAF et al., 2003). DPGrams of all subjects showed measurable DPOAE levels in the frequency range between 1 and 6 kHz (ILO92, Otodynamics Ltd., UK). Recordings were performed in that ear (right or left) in which the DPOAE level at $f_2 = 2$ kHz was higher.

Measurements were performed in a sound-insulated booth while subjects were seated in a comfortable recliner and were encouraged to read a book.

2.2. Equipment and signal analysis

Acoustic signal generation and recording were done using a 24 bit soundcard (GINA, Echo Corp., USA) within a Pentium computer, and MATLAB customized software. The sound probe for primary tone generation and DPOAE recording was an ER-10C (Etymotic-Research, USA). The primary tone frequencies were set to $f_1 = 1.67$ and $f_1 = 2.0$ kHz, the levels to $L_1 = 51$ and $L_2 = 30$ dB SPL to yield optimal conditions for eliciting DPOAEs of $f_{DP} = 2f_1 - f_2$. The reason for using relatively low-level primary stimuli is the fact that the resulting DPOAEs are more sensitive for minute changes in cochlear micro-mechanics than high-level DPOAEs (Kummer et al., 2000; Janssen, 2005). The acoustic parameters of the primary tones were kept constant in all measurements.
The biasing tones were generated using a modified headphone (Beyerdynamic DT-48 earphone, Germany), from which the sound was delivered by a thin polythene tube (1 mm diameter, 20 cm length) into the outer ear canal through the earplug of the ER-10C sound probe (Fig. 1). The IF frequencies of the biasing tone were $f_B = 6$ and 12 Hz. Additionally, biasing tones within the LF range of 24 and 50 Hz were used for comparison. The soundcard’s output was power boosted and low-pass filtered (6 dB/octave, $f_g = 60$ Hz). The levels of biasing tones were adjustable between 80 and 130 dB in 1 dB steps.

DPOAEs were analyzed in the frequency domain by Fast Fourier Transform (FFT) of the averaged time-domain waveforms (8 epochs of 5 s each). The right panel of Fig. 2 shows the important part of the spectrum of the modulated DPOAE signal, the carrier at $f_{DP} = 2f_1 - f_2$ and the spectral components of the modulation spaced by $f_B$ either side of the carrier. Only these five spectral components were used to compute the time course of the modulated DPOAEs by inverse FFT. As it turned out, the inclusion of higher order spectral side lines did only added noise to the analysis. The envelope of this signal represents the periodical time course of the modulated DPOAE amplitude and is a function of the running phase $\phi_B$ of the biasing tone, and referred as modulation period pattern (bottom right, two periods shown).

The left panel of Fig. 2 shows the spectrum when no biasing tone is applied. Since the time course of the DPOAEs was computed in the same way the poor fluctuations in the time course are mere noise artefacts due to the noise floor present at the sideline frequencies. This gives a graphical indication of the impact of noise inherent in the modulated time course given in the right column of Fig. 2.

To estimate the measurement quality, two types of narrow-band signal-to-noise ratios (SNR) were defined in the frequency domain. First, the so called carrier–to-noise ratio (CNR), which describes the quality of measurement for the unmodulated DPOAEs was calculated referring to the
RMS mean noise power level within a small spectral band around $f_{DP}$ using only frequencies between the modulation sidebands. DPOAEs without a biasing tone were accepted when CNR $\geq 6$ dB. Second, the so-called modulation-to-noise ratio (MNR) described the relation between the modulation sidebands, without the carrier line, and the noise, describing only the modulation power vs. noise. Modulated DPOAEs were accepted, when CNR and MNR $> 6$ dB. The definition of MNR was crucial, because the conventional SNR, which includes carrier and modulation sidebands, would not indicate whether or not the observed modulation was only due to noise.

### 2.3. Calibration

The calibrations were performed using an occluded-ear simulator (B&K4157, Brüel & Kjær, Denmark; according to IEC 711: 2006). Since the frequency response of the occluded-ear simulator’s output phase in the frequency range below 100 Hz is neither defined by IEC 711 nor by the manufacturer, it had to be calibrated before all other calibrations. This was done only once and is referred to as *boot calibration*. In a second step the microphone of the ER-10C was calibrated in the occluded-ear simulator. This procedure was repeated whenever a new foam probe was placed in the ear canal, the sound probe’s receivers and the modified earphone (DT-48 with tube) were calibrated to account for the individual ear canal volume. This is called *operational calibration*. After *boot calibration* and *operational calibration*, when the probe was placed in the ear canal, the sound probe’s receivers and the modified earphone (DT-48 with tube) were calibrated to account for the individual ear canal volume. This is called *probe check*, and done frequently during all sessions for monitoring and readjusting the probe fit in the ear canal.

For *boot calibration*, the complex frequency response $S_{ref}$ of the occluded-ear simulator’s microphone was compared to the respective complex frequency response $S_{HF}$ of a HF-carrier microphone system (B&K2631). This system has a lower limiting frequency of 0.01 Hz and has virtually no linear phase or amplitude distortion at the lowest biasing frequency used, namely 6 Hz. The HF carrier microphone’s standard ½ in. diameter sound inlet was coupled face-to-face to the occluded ear simulator’s orifice, also approx. ½ in. wide, by means of a short silicone rubber tube (inner diameter 12.5 mm, length 20 mm). The volume within this tube was fed with test sound generated by a modified DT-48 headphone similar to the device used for the DPOAE measurements, via a polythene tube (inner diameter 1 mm, length 30 mm). Venting of the volume was done via a 0.1-mm-diameter flexible PVC tube of 100 mm length. In order to get a sufficient SNR in the IF/LF-range, especially for phase measurements, a series of discrete test tones was applied (steps of 1.5 Hz for frequencies between 1.5 and 15 Hz and steps of 3 Hz between 15 and 24 Hz). Values at intermediate frequency points were calculated by linear interpolation. For frequencies above 24 Hz, no differences between B&K4157 and B&K2631 in amplitude or phase were found.

For *operational calibration*, the complex frequency response $S_{probe}$ of the ER-10C microphone were measured with broadband pseudo-random noise of 1.5 Hz–12 kHz, produced by the DT-48 (1.5–200 Hz) and the ER-10C probe receivers (200 Hz–12 kHz), using FFT for conversion to the frequency domain. The numerical reference values $S_{ref}$ obtained during the boot calibration were used as a complex spectral weighting function compensating for the frequency response errors of the simulator’s microphone output path. The inverse of the obtained frequency response $S_{probe}$ could now be applied as a complex weighting to all spectra recorded with the ER-10C microphone.

For measurements in the ear canal, this calibration of the ER-10C microphone was regarded as sufficiently precise only for frequencies below the range where the impedance of the individual ear canal starts to deviate from the impedance of the occluded-ear simulator, i.e., where standing waves above 4 kHz occur.

During *probe check*, the complex frequency responses of the DT-48 and of the two ER-10C receivers were obtained in the individual ear canals. The same stimulus as in the operational calibration was applied to these receivers, and the sound pressure in the ear canal was recorded via the ER-10C microphone. Due to the restrictions at operational calibration, the probe check is valid for frequencies below approx. 4 kHz. The deviation of the impedance between the occluded-ear simulator and the individual ear canal can be neglected because primary tones were used only, which were well below the frequency range where standing waves occur in the ear canal. Like for *operational calibration*, the three stimulus transducers produced periodic noise from 1.5 to 200 Hz (DT-48) and 200 Hz to 12 kHz (ER-10C receivers). The ratio of the respective spectra of the electrical input signal and the spectra of the sound pressure yielded complex spectral weighting functions for compensation of the linear amplitude and phase distortions of the stimulus transducers during DPOAE measurements.

#### 2.4. Harmonic distortions

The harmonic distortion of the DT-48 was monitored by means of the overall FFT during sound calibration in the occluded-ear simulator and during DPOAE measurements in the ear canal. Fig. 3 shows the fundamental and the harmonics of the biasing tone $f_B = 6$ Hz measured in the occluded-ear simulator. All harmonics of this biasing tone lay below both the hearing threshold (ISO 226: 2003) and the estimated infrasonic threshold (Møller and Pedersen, 2004).

To avoid mutual influence, the primary tone frequencies were chosen so that $f_{DP}$ and its modulation sidebands did not coincide with any higher order harmonics of the biasing tones or any modulation sidebands of the primaries.
2.5. Definition of modulation index

For characterizing DPOAE modulation patterns by a single scalar value we did not use the modulation depth (MD) (see Fig. 4a) or the modulation percentage (m), known from AM-broadcast theory, since the modulation of the DPOAEs (2f1 − f2) by the biasing tone is not a pure amplitude modulation, but contains a certain amount of phase modulation depending on the subject (Scholz et al., 1999). Rather we devised a new measure using the time-variant complex vector of the modulated DPOAEs in polar coordinates (Fig. 4b). This measure, called the modulation index (MI), combines both amplitude and phase changes, and will be described below.

Note that there are two different phase-time-functions: One is the time-variant zerophase angle \( \varphi_{\text{DP}} \) of the modulated DPOAE itself (as compared to a hypothetical steady-state vibration of this frequency), which is the angle in the polar pattern in Fig. 4b. The other is the running phase angle \( \varphi_B \) of the sinusoidal biasing tone (see Fig. 4a). The locus curve in Fig. 4b describes the DPOAEs during one period of the biasing tone, and the reference numbers in the time domain (Fig. 4a) correspond to the characteristic points on the locus curve (Fig. 4b). The length of the radius

![Fig. 4](image_url)

Fig. 4. Modulated DPOAE using a biasing tone of \( f_B = 6 \text{ Hz} \) and \( L_B = 130 \text{ dB SPL} \). (a) Magnitude modulation pattern with modulation depth MD and biasing tone below. (b) Locus of complex amplitude: The vector travels once per biasing period, passing through time instants 1–4 (indicated in a). Modulation span MS, and stationary part (carrier) DPSTAT of the DPOAE are indicated.
vector to any point on the curve marks the instantaneous magnitude of the complex amplitude. Its minimum and maximum defined the previously used MD as shown in Fig. 4a. The angle between the radius vector and the positive real axis describes the instantaneous zero-phase angle $\varphi_{pp}$. This diagram makes it quite obvious that the phase contribution needs to be considered when quantifying the DPOAE modulation. The phase value $\varphi_{\text{ref}}$ of the pronounced minimum (reference number 4 in Fig. 4) in the modulation pattern is taken as a characteristic value of the modulation pattern along the $\varphi_{\text{ref}}$ phase axis. Consequently, we characterize the modulation by the maximum width of the locus curve, shown graphically by the maximum diameter, here called modulation span (MS). The radius vector to the centre of that line approximately represents the stationary part $\text{DP}_{\text{stat}}$ of the modulated DPOAE, i.e. the sinusoidal carrier of the modulation (Fig. 4b). Thus, the special modulation index MI is defined by half the modulation span normalized by the stationary part ($\text{DP}_{\text{stat}}$): $\text{MI} = 1/2\text{MS}/\text{DP}_{\text{stat}}$.

Assuming that the cochlear partition is displaced between scala vestibuli and scala tympani by the biasing tone with the periodicity of $f_B$, the operating point of the cochlear amplifier changes also with this periodicity. Consequently, the magnitudes and phases of the DPOAEs vary with time and show a periodical pattern. The individual value of MI depends on one hand on the transfer function of the sound pressure at the eardrum to the displacement of the cochlear partition, and on the other hand on the transfer function between displacement of cochlear partition and DP modulation strength.

2.6. Sequence of experiments

After the selection of the ‘better ear’ (see Section 2.1., Subjects) and after the probe check, the first DPOAE recording was done without a biasing tone in order to get an individual initial value for each subject. Then followed the recording of DPOAEs modulated by an IF tone of $f_B = 6$ Hz with $L_B = 130$ dB SPL. The subjects were free to immediately switch off the acoustic stimuli by pushing a button if they felt any annoyance. After this first recording of 40 s duration the subjects were asked how they felt and their eye movements were visually inspected for nystagmus. None of the subjects reported any annoyance or dizziness, or showed signs of nystagmus, in which case all further tests would have been abandoned.

The calculated MI of the first recording with a 6 Hz biasing tone was taken as an individual reference value for subsequent measurements using biasing tone frequencies of 12, 24, and 50 Hz. Here the biasing tone level was adjusted in 1 dB steps until the MI was within ±10% of the individual reference value. Finally, DPOAEs were recorded once more without a biasing tone for comparing DPOAEs before and after the biasing experiments. The difference of the DPOAE amplitudes of both recordings were used to indicate a possible noise induced temporary threshold shift (TTS) which might occur during IF or LF tone stimulation as suggested by Harding and Bohne (2004). They also used the recorded DPOAE level-shift as best alignment with detailed pathology on noise-induced cochlear damage in animals (Harding et al., 2007).

On average, nine recordings per subject were necessary to test all four biasing tone frequencies (40 s each). A pause of 1 min was set between recordings. Altogether, the recordings took about 30 min per subject. Immediately after the last test, the subjects were again asked about any feeling of dizziness, annoyance or hearing problems, and were again tested for a nystagmus reaction. This interview was repeated after further 10 minutes.

3. Results

3.1. Individual data

Fig. 5 plots the DPOAE modulation patterns for different biasing tones (dashed lines) in the right ear of subject L.H.; for comparison DPOAE amplitude is shown without a biasing tone (solid line). DPOAE amplitude (Fig. 5a) is drawn as a function of the phase of the respective biasing tones (two periods each, see Fig. 5b). Here the DPOAE amplitude without a biasing tone was 13 $\mu$Pa with small deviations due to noise. The maximum deviation during recording, individually depending on the actual SNR, was about 1 $\mu$Pa. When the biasing tones were present, DPOAE amplitudes changed considerably.

A biasing tone of 6 Hz and 130 dB SPL modulated the amplitude between 6 and 16 $\mu$Pa (short-dashed line). A distinct minimum occurred near $\varphi_{\text{ref}} = 270^\circ$, where the cochlear partition is supposedly being displaced towards scala tympani. With increasing frequencies of the biasing tone $f_B$, the minimum DPOAE amplitude showed increasing phase-lags $\varphi_{\text{ref}}$. At $f_B = 50$ Hz the phase-lag amounted to almost $130^\circ$ (re 6 Hz). DPOAE modulation patterns were quite similar for both IF and LF tones, while an almost constant MI was maintained in the sequence of experiments, as described above.

A temporary enhancement of the DPOAE amplitude above the unbiased DPOAE level (here 13 $\mu$Pa, solid line) was observed within all recorded modulation pattern, however to a different degree. This enhancement of the modulation pattern is a feature of the modulation itself and was found to be individually pronounced.

3.2. Pooled data

The modulation patterns obtained in the other subjects resembled those shown in Fig. 5a, exhibiting only small individual differences in MI and MD, and higher differences in phase $\varphi_{\text{ref}}$. The phase-lag of the DPOAE minima increased with increasing $f_B$ also in the pooled data. The individual modulation pattern was well robust but the intersubject phase-lags varied considerably. As shown in Table 1 the median of phase-lag $\varphi_{\text{ref}}$ amounted to $-139^\circ$. 

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### Table 1

<table>
<thead>
<tr>
<th>Frequency (Hz)</th>
<th>Phase Lag ($^\circ$)</th>
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<tbody>
<tr>
<td>6</td>
<td>-139</td>
</tr>
<tr>
<td>12</td>
<td>-110</td>
</tr>
<tr>
<td>24</td>
<td>-100</td>
</tr>
<tr>
<td>50</td>
<td>-90</td>
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at $f_B = 50$ Hz, referred to 6 Hz. The quartiles also increased with increasing $f_B$.

The median of MI of all subjects as a parameter for individually constant modulation was $MI = 0.44$, the quartiles $q_{25}$ and $q_{75}$ were 0.35 and 0.64, respectively. For comparison, the modulation artefact of the DPOAEs without a biasing tone, shown in Fig. 5a, has a $MI = 0.046$.

Table 2 shows the medians and quartiles ($q_{25}$ and $q_{75}$) of the sound pressure level $L_B$ of the biasing tones for yielding an individual constant modulation index (MI) (pooled data).

<table>
<thead>
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<th>$f_B$ (Hz)</th>
<th>$L_B$ (dB SPL)</th>
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<tr>
<td></td>
<td>$q_{25}$</td>
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<tr>
<td>6</td>
<td>130.00</td>
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<tr>
<td>12</td>
<td>114.75</td>
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<td>24</td>
<td>100.00</td>
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<td>50</td>
<td>90.25</td>
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Table 2

Median with 25%, and 75% quartiles ($q_{25}$, $q_{75}$) of the sound pressure level $L_B$ of the biasing tones for yielding an individual constant modulation index (MI) (pooled data).

<table>
<thead>
<tr>
<th>$f_B$ (Hz)</th>
<th>$L_B$ (dB SPL)</th>
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<td>50</td>
<td>90.25</td>
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Table 1

Median with 25% and 75% quartiles ($q_{25}$, $q_{75}$) of the phase lags $\varphi_B$ ($f_B$) in degrees related to the value $\varphi_B$ of the 6 Hz biasing tone for yielding an individually constant MI (pooled data).

<table>
<thead>
<tr>
<th>$f_B$ (Hz)</th>
<th>$\varphi_B$ ($^\circ$)</th>
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<tbody>
<tr>
<td></td>
<td>$q_{25}$</td>
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<tr>
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<td>−20</td>
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<td>24</td>
<td>−48</td>
</tr>
<tr>
<td>50</td>
<td>−117</td>
</tr>
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</table>

Table 1

Median with 25% and 75% quartiles ($q_{25}$, $q_{75}$) of the phase lags $\varphi_B$ ($f_B$) in degrees related to the value $\varphi_B$ of the 6 Hz biasing tone for yielding an individually constant MI (pooled data).

at $f_B = 50$ Hz, referred to 6 Hz. The quartiles also increased with increasing $f_B$.

The median of MI of all subjects as a parameter for individually constant modulation was $MI = 0.44$, the quartiles $q_{25}$ and $q_{75}$ were 0.35 and 0.64, respectively. For comparison, the modulation artefact of the DPOAEs without a biasing tone, shown in Fig. 5a, has a $MI = 0.046$.

Table 2 shows the medians and quartiles ($q_{25}$ and $q_{75}$) of the biasing tone level $L_B$ that were required to obtain a MI close to the individual reference value, established for $f_B = 6$ Hz and $L_B = 130$ dB SPL. For all subjects the necessary level $L_B$ of the biasing tone for yielding constant MI decreased with increasing $f_B$. Note that the deviation from the medians was also very small.

The median values and quartiles of $L_B$ (see Table 2) were plotted in Fig. 6 (solid line) together with the hearing threshold (ISO 226: 2003, dashed line) and the IF threshold (dotted line). The interpolated function of the medians of $L_B$ shows a slope of $-13$ dB/octave between 6 and 12 Hz, $-15$ dB/octave between 12 and 24 Hz, nearly parallel to the estimated detection threshold in the IF range, and a slope of $-9$ dB/octave in the LF range between 24 and 50 Hz.

None of the subjects in the study group pressed the button to stop a measurement. Each subject described the IF biasing tones as a sensation of vibration at the eardrum, not as a tonal audible stimulus. In addition, most subjects reported a weak but clearly audible sound sensation, described as humming. Immediately after the last recording none of the subjects showed a nystagmus reaction. There were no significant differences between the levels of DPOAEs at $f_2 = 2$ kHz without biasing tones before and after the measurements, indicating that at least near the 2 kHz
region the experimental exposure by the biasing tone did not affect cochlear integrity. In the interviews 10 min after the last auditory stimulus, none reported any annoyance, aural pain, tinnitus, stress or dizziness.

4. Discussion

Employing the modulation of DPOAEs with biasing tones of 6 and 12 Hz, our results show that infrasound obviously affects the human inner ear function. The impact of the fundamental of the infrasound biasing tones is clearly evident in the DPOAE modulation patterns. That means that infrasound itself has an impact on the cochlear partition, not only by means of harmonics above 16 Hz.

Similar modulation patterns in humans were shown psychophysically by Zwicker (1977) and Mrowinski et al. (1995). In normally hearing subjects they measured pronounced modulation patterns of masked hearing thresholds of clicks and short tone bursts that depended on the running phase of the biasing tone. Later, Frank and Kössl (1996, 1997), Bian et al. (2002, 2004) and Lukashkin and Russel (2005) showed in animals that a loud low-frequency tone can modulate the amplitudes of DPOAEs. The modulation of the hearing threshold became more dominant with increasing $L_B$. This and other parameter dependencies were also studied by Scholz et al. (1999) for LF modulated DPOAEs in human subjects using a constant biasing frequency of $f_B = 32$ Hz.

Previous studies that analysed DPOAE modulation patterns used MD as quantitative single-value parameter to describe the magnitude modulation patterns, and none of the OAE biasing studies considered the coexisting phase modulation. In the present study, MI was calculated to describe the amount of modulation, comprising both MD and variation of phase $\phi_{DP}$.

There are at least two possible reasons underlying the observed intersubject differences in MI during the initial 6 Hz biasing tone exposure of 130 dB SPL: First, the gain of the middle ear transfer function might vary and lead to different displacement amplitudes of the cochlear partition in the individual ears. Second, it is possible that the shapes of the gain functions of the individual cochlear amplifiers were different, leading to variations in shape and size of the resulting DPOAE modulation patterns. In the following, we focus our discussion on the middle ear transfer function influenced by the cochlear impedance.

The present study was designed to compare the modulation at different biasing tone frequencies, especially between IF and LF sounds. The results show that in order to keep the modulation strength constant the biasing tone levels at different biasing frequencies describe a level linear function of $-12$ dB/octave slope, regardless whether the biasing tone frequencies lie in the IF or LF region.

In Zwicker’s masking experiments (1977) of a short tone burst using LF masking sounds, a shift in threshold and pitch was observed which was linked to the first derivative of the sound pressure in the ear canal for masker frequencies above 40 Hz, and linked to the second derivative for masker frequencies below 40 Hz. The latter indicates a middle ear transfer function with a slope of 12 dB/octave in this frequency range. The iso-modulation-curve in this study, representing the inverse of this transfer function, showed the same characteristic: The overall slope of the
function of the biasing levels resulting in equal MI between 6 and 50 Hz was approx. −12 dB/octave (Fig. 6, bold curve). Also the 180° phase lead of the maximum DPOAE suppression (ϕ_m) compared to the ear canal pressure (peak value), seen at the lowest modulation frequency of 6 Hz, supports this notion. The apparent reduction of this phase lead (Table 1) with increasing f_b is mostly accountable for by a constant reverse travelling time of the DPOAE, which is at the order of a couple of milliseconds, and translates to a larger phase shifts as f_b increases.

Middle ear transfer-functions at low frequencies have been obtained in four animal species by Dallos (1970). Similar to our assumption that equal MI indicates equal displacement of the cochlear partition, in his study the amplitude of the recorded cochlear microphonic signal is assumed to be proportional to the cochlear partition displacement. Like in our experiment, Dallos kept the amplitude of the cochlear microphonics constant for every frequency of the LF stimulus by adjusting the sound pressure in the ear canal. Thus, the meaning of the resulting curves is very similar to the iso-modulation-curves of the present study. In the relevant frequency range between 20 and 50 Hz, Dallos found slopes of −12 dB/octave for cat and chinchilla, and −6 dB/octave for guinea pig and kangaroo rat. Based on mechanical model calculations, Dallos suggested, that the slope in this frequency region was linked to the balance between inertia and viscous resistance associated with the flow of perilymph through the helicotrema: Animals with wide helicotremae and little tapering of the cochlear ducts (cat, chinchilla) should have −12 dB/octave below the transition frequency, while animals with a narrow apical cochleae (guinea pig, kangaroo rat) should have −6 dB/octave slopes. Based on anatomical similarities with cat and chinchilla cochleae, he predicted for the human cochlea a −12 dB/octave slope. Our results confirm his conjecture.

Although probably to a lesser extent, these anatomical variations seen between species could also exist between individual ears. Thus, in addition to variations in the gain of the middle ear transfer function, individual differences in the slope of the middle ear transfer function might be an explanation for the wide range in observed sensitivity and tolerance to IF and LF sound (Feldmann and Pitten, 2004).

Based on the hypothesis that the influence of the perilymph inertia decreases with −12 dB/octave towards lower frequencies, and the influence of the viscous resistance decreases only with −6 dB/octave, Dallos postulated that at frequencies below this cross-over the slope of the middle ear transfer-function will change to −6 dB/octave in accordance with a viscosity dominance. Our data show that at 6 Hz this condition is not yet reached in the human cochlea.

There exists no international standard for the threshold in the infrasonic frequency range. Møller and Pedersen (2004) reviewed published data on IF/LF hearing thresholds and equal loudness contours which were measured using a variety of different psycho-acoustic methods under differing conditions. They fitted a first order linear regression through the available IF data which is plotted for reference in Fig. 3. Its slope is close to that of the iso-modulation curve which is obtained with much higher IF levels. The LF hearing threshold curve (ISO 226: 2003), on the other hand, is considerably steeper than our median iso-modulation curve. The 90 phon equal loudness curve, however, which corresponds to the sound pressure levels applied in our experiments, has less slope, and is close to the slope of the iso-modulation curve. These findings are consistent with equal loudness contours, which show less level-dependent change in slope at IF frequencies compared to the LFs (Møller and Pedersen, 2004).

Regarding the annoyance of IF/LF noise, Leventhall et al. (2003), Leventhall (2004) and many others have stated that present limits on environmental noise level, based on A-weighted filters, are inadequate. Equal loudness contours at lowest frequencies become more dense, and thus it seems obvious that loudness perception and consequently annoyance increases dramatically in this frequency range. The results of our study showed that the displacement of cochlear partition, caused by an IF tone of 6 Hz and 130 dB SPL, clearly affected the auditory processing in the human cochlea. The absence of temporary DPOAE level-shift (TDLS) indicates that the stimuli did not cause damage to the OHCs or their hairbundles, at least at the DPOAE generation site. Harding and Bohne (2004) and Harding et al. (2007) showed that TDLS was a more sensitive measure than TTS, normally used to look for damages of OHC after noise exposure.

Van den Berg (2001) found that 10% of normally hearing subjects aged from 50 to 60 years reported annoyance during extended exposure to high level LF noise. In the present study none of the subjects complained about annoyance during or after the biasing tests, probably because of their positive attitude to the experiments, or because the exposure was only of short duration. In a research project for the American Space Program, Mohr et al. (1965) exposed subjects to tones at 10 and 20 Hz at levels of 150 dB SPL for 2 min and found no measurable TTS one hour later. However, Jerger et al. (1966) reported on TTS between 3 and 6 kHz in subjects after they had been exposed for 3 min to infrasound tones with frequencies between 2 and 12 Hz and levels of 120–140 dB SPL. Hearing recovered completely after one hour. It is however known that infrasound affects the human organism. Details about the occurrence, the character and the psychological aspects of the so-called hum phenomenon (complains about LF environmental noise, which is not normally detectable by sensitive measuring equipments) were described by Leventhall (2004).

An explanation for the individual variation in sensitivity or and tolerance to infrasound exposure requires further investigation. The presented method of measuring noninvasively the IF/LF response in human cochlea might become valuable for this research.


