Laser interferometric vibration measurements of the middle ear in healthy humans

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ABSTRACT

The use of spontaneous and evoked otoacoustic emissions is now a standard clinical tool for diagnosis of the function of the inner ear. However, it is not possible to extract this information over the entire, functionally relevant frequency range because of imperfect coupling of: i) stapedial to ear-drum vibrations through the ossicular chain of the middle ear and ii) ear-drum vibrations to air in the external auditory meatus. The problem could be circumvented if it were possible to measure the vibration of the stapes and ear drum. The ear drum can be visualized non-invasively, whereas the stapes is only accessible intra-operatively. Therefore, we designed a laser-interferometric system to non-invasively measure the vibration of the human ear drum.

Vibrations were measured with a laser Doppler velocimeter (Polytec OFV-302) coupled into the side arm of an operating microscope (Zeiss OPMI MDM). The wavelength was 633 nm and emitted power was less than 1 mW. Direct coupling through the optics of the operating microscope, instead of through glass fibres, enabled a larger signal-to-noise ratio (20-30 dB) due to collection of more reflected light. This coupling scheme avoids the problems associated with having to place a reflecting material on the ear drum.

The developed vibration measurement system allows non-invasive, fast and reproducible characterization of the dynamics of the human ear drum and as such can be used for clinical diagnostics.

Keywords: Laser Doppler velocimeter, hearing, human ear

1. INTRODUCTION

There is often a plethora of diseases in the middle ear cavity, but unfortunately, this region of the ear is visually inaccessible. Fig. 1 shows the anatomy of the middle ear. One can observe how the eardrum (tympanic membrane) hermetically closes off the middle ear from the ear canal (external auditory meatus), permitting no access from the outside. Consequently only indirect measurements of middle-ear function can be made through the ear canal. The proposed measurement technique should produce sufficient information about the function of the middle ear and possibly also the inner ear to allow reliable differential diagnoses.

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With present clinical methods, the diagnostic differentiation between different middle-ear pathologies (otosclerosis, ossicular interruption, adhesive otitis, etc.) is very difficult. For example, the most important of these methods is conventional audiometry. This procedure measures the minimal subjective auditory threshold, using air- and bone-conducted sounds. The air conduction test provides information on the sensitivity of the whole hearing system. The bone conduction test, by-passing the middle ear, is supposed to measure losses in the inner ear only. A difference in the curves obtained from these two tests reveals a loss due to middle-ear diseases. Unfortunately, this procedure is not always reliable because of uncertainties inherent to the bone conduction test, which is often influenced by middle-ear disease. Another problem is that the function of the inner ear cannot be measured above 10 kHz with these tests. There are other methods such as impedance, stapes reflex, etc., but these are used complementarily and alone are not sufficient for a diagnosis.

Vibration measurements of the human tympanic membrane *in vivo* have been published with holography[7,5] and laser-Doppler velocimetry[8]. The main disadvantages of these techniques for clinical purposes are: i) long measurement time; ii) need for reflecting materials to be placed on the tympanic membrane; iii) introduction of glass-fibre optics into the ear canal.

Here we describe a method for *in vivo* measurement of middle-ear dynamics which is eminently suitable for clinical purposes.

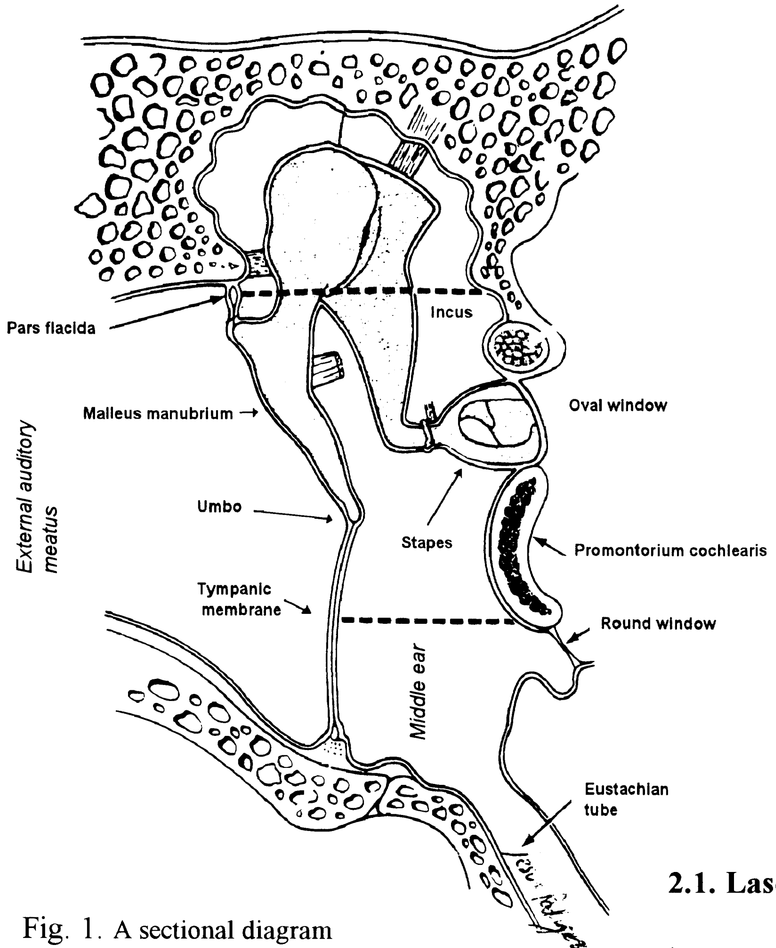


Fig. 1. A sectional diagram of the ear.

2. MATERIALS AND METHODS

2.1. Laser interferometry

A commercial laser Doppler vibrometer (Model OFV-3001 with an OFV- 302 sensor head, Polytec) was used to make the measurements of tympanic membrane motion. The laser was coupled to a standard operations microscope (OMPI MDM, ZEISS) with a 200 mm front lens (Fig. 2), equipped with a continuous zoom with magnification factors from 0.6 to 2.4. The total power in the focused (diameter: 72 μm) laser beam was 0.75 mW. The amount of reflected light was sufficiently large to avoid having to place reflective materials on the tympanic membrane. The use of an operations microscope, instead of glass fibre optics [8,9], enabled non-invasive measurement of tympanic-membrane motion.

2.2. Acoustic stimulation

Sound was delivered via a plastic ear speculum (otoscope) inserted into the external auditory canal. Sound pressure was monitored within 4 mm of the tympanic membrane, using a probe microphone (B&K 4182) and measuring amplifier (B&K 2610), and compensated for resonances in the sound delivery tube. Measurements were made at 60-70 dB SPL.

2.3. Frequency response measurement technique

Since the middle ear vibrations were linear (distortion 40 dB below fundamental at 100 dB SPL), broad band stimuli were used, which allow the determination of frequency response functions over a wide frequency range with only one measurement. We used bandlimited white noise (50 Hz - 20 kHz, 1024 sample points repeated every 20 ms), which was produced by the signal generator of a frequency analyzer (A&D 3525). To reduce the energy delivered to the loudspeaker, but to obtain a high signal-to-noise ratio, a multi-tone complex was used. It consisted of 130 spectral points (50-19550 Hz), all harmonics of 50 Hz but with a relative distance of at least 1.026 (27.3 points per octave). The amplitudes were equal but the phases random. The energy of this signal was a factor of 3.1 less than the white noise signal. In the frequency range from 10 kHz - 20 kHz, where the output of the sound delivery system was low, a multi-tone complex from 10-20 kHz (28 spectral points, logarithmically spaced with 27.3 points per octave) was used to provide sufficient energy for a good signal-to-noise ratio. The energy of this signal was a factor of 14.3 less than for the white noise stimulus.

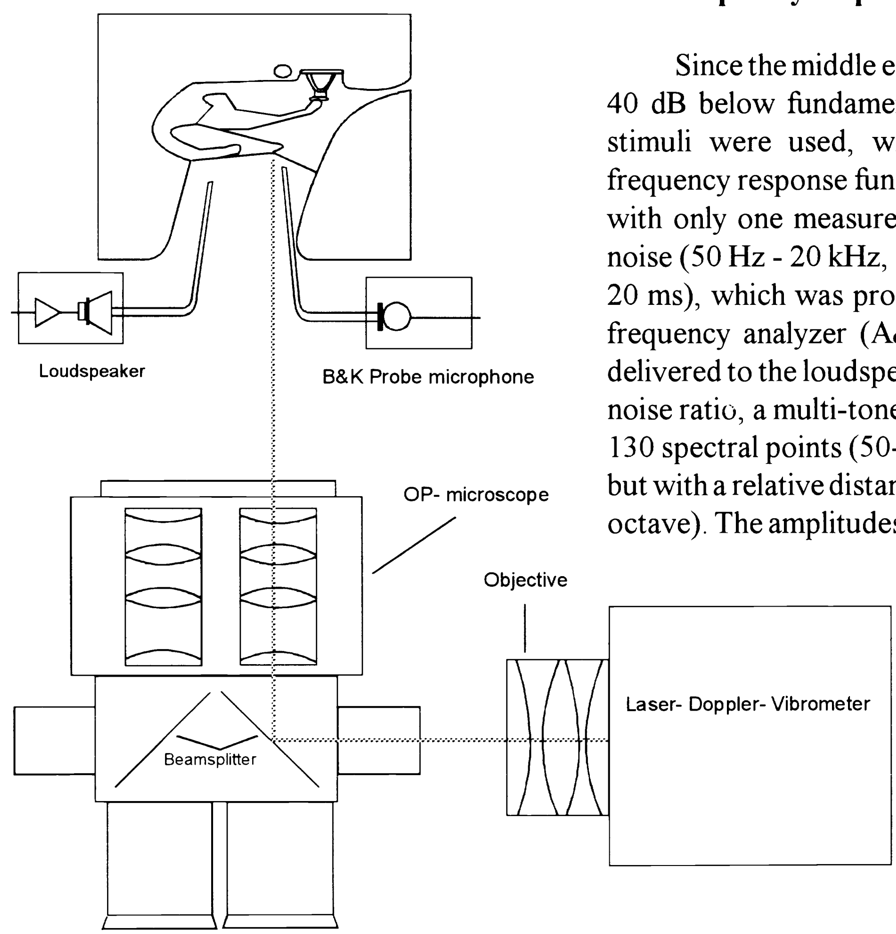


Fig. 2. Schematic diagram of the measurement system for the tympanic membrane.

The stimuli were synchronized with the analysis frame. The amplitude and phase responses of the tympanic membrane were calculated as the Fourier transform of the output signal divided by the Fourier transform of the input signal. Averaging in the time domain (50 frames) allowed rapid determination of the frequency response. The measurement time was 20 s.

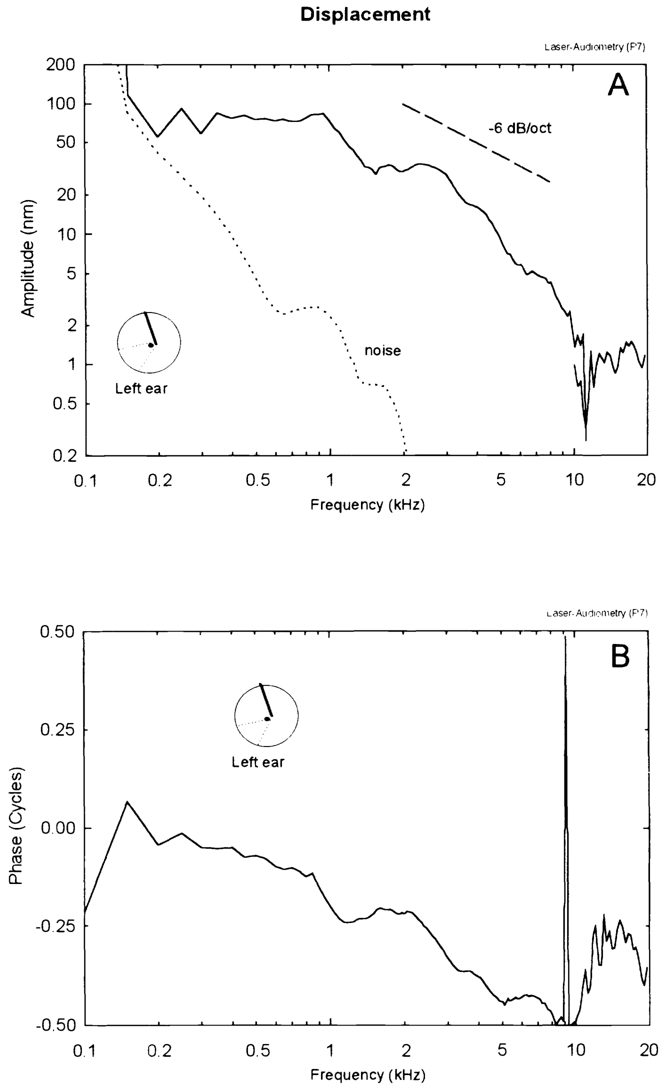


Fig. 3. Amplitude (A) and phase (B) of umbo displacement for 100 dB SPL near the the tympanic membrane.

The measurements were performed on five volunteer subjects after completing a screening process to ensure that the experimental sample consisted of normal hearing ears. The ear canal and drum were inspected otoscopically. The following experiments were performed:
a Békésy audiogram from 10 Hz to 20 kHz, to establish hearing thresholds, a tympanogram, stapedius reflex test, as well as transient otoacoustic emissions (ILO88) and brainstem evoked response audiometry to exclude cochlear or retrocochlear pathology. The subject lay supine on a bench. The external auditory canal was straightened and widened by the insertion of the customized otoscope. This gave optical access to a large area of the tympanic membrane. The probe microphone was introduced and placed near the drum and the distance measured and adjusted to be less than 4 mm. Sound pressure was then measured and equalized.

Data acquisition was performed in two frequency ranges: 50 Hz to 20 kHz and 10 to 20 kHz. The latter frequency range ensured adequate signal-to-noise above 10 kHz.

Measurement at different points on the membrane were performed in three subjects. For one subject, pressure and vibration data were collected for the probe tube at different distances from the tympanic membrane and for another subject data were collected on two consecutive days.

3. RESULTS

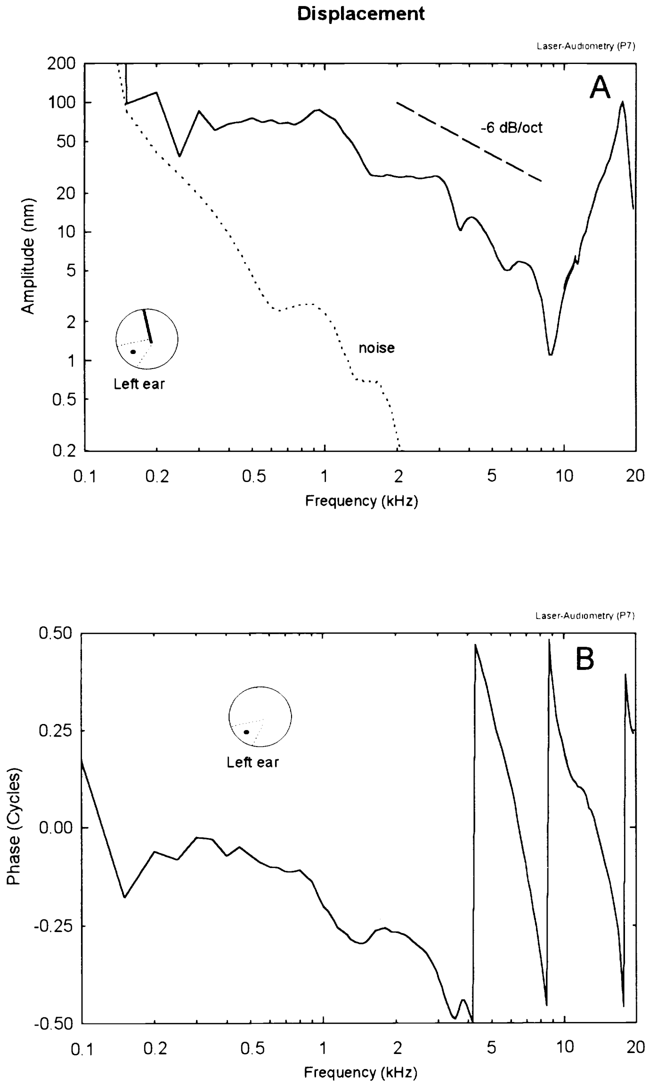


Fig. 4. Amplitude (A) and phase (B) of tympanic membrane displacement for 100 dB SPL. Same subject as in Fig. 3.

The displacement response, corrected to 100 dB SPL, is shown in Fig. 3 for one subject. At frequencies between 0.1 and 1 kHz the amplitude response on the umbo is relatively flat. The amplitude of the vibration is approximately 100 nm / 100 dB SPL. At frequencies between 1 kHz and 5 kHz the displacement decreases at approximately 6 dB per octave and is accompanied by small undulations. Data around 11 kHz are limited by poor signal-to-noise ratio.

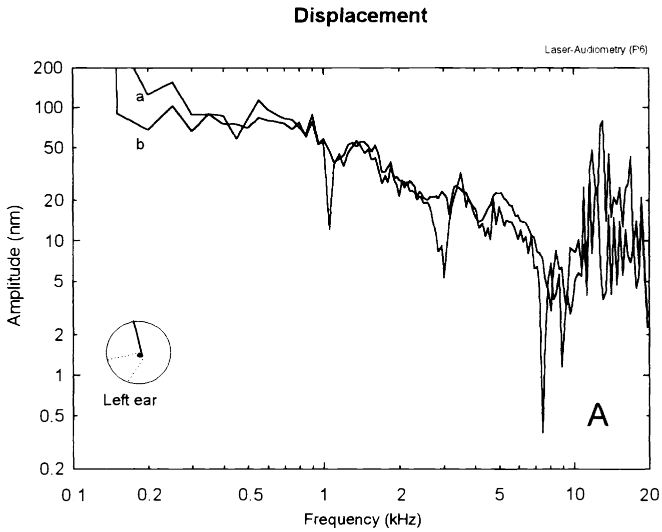
Two resonances are apparent between 1 kHz and 10 kHz: the first between 900-1200 Hz and the other at about 3 kHz. Although these resonances were apparent in other subjects, they were by no means of the same magnitude in each subject.

Fig. 4 A shows responses obtained from another point on the drum. A deep antiresonance is seen at 9 kHz. There are also antiresonances at 3.5 kHz and 6 kHz. The antiresonances were obtained in all subjects.

In all cases the phase responses were consistent with the amplitude responses (in a minimum-phase sense).

3.1. Repeatability of the method

Fig. 5 A shows results from the same subject measured 24 hours apart. The results are similar, apart from deviations at 1 and 3 kHz which were due to the shorter averaging time in *a*.



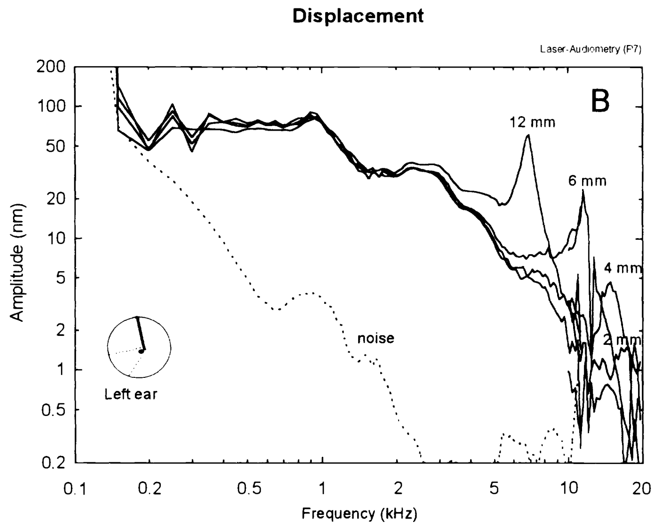


Fig. 5. A: Comparison of amplitude responses of the umbo measured 24 hours apart (b: after 24 hours).

B: Comparison of amplitude responses of the umbo for the probe microphone at different distances from the tympanic membrane.

3.2. Standing waves

In order to determine the maximal allowable distance of the probe microphone from the tympanic membrane before interference from standing waves becomes appreciable, measurements of sound pressure and displacement were made for different probe-microphone distances. Referring to Fig. 5 B, the resonances at 6.5 kHz and 13 kHz for distances of 12 mm and 6 mm, respectively, correspond to the quarter-wavelength resonances. In general, displacement data are independent of frequency up to 10 kHz for distances less than 4 mm and independent of frequency up to 6 kHz for distances less than 6 mm. Consequently, the distance was routinely chosen to be less than 4 mm.

3.3. Specific acoustic impedance

The specific acoustic impedance of the middle ear, defined here as the ratio of sound pressure near the tympanic membrane to velocity of the umbo, was minimal in the region between 1 kHz and 7 kHz (Fig. 6). The impedance amplitude increased to lower and higher frequencies, with the phase tending to -0.25 cycles and 0.25 cycles, respectively. The form of the impedance curves was similar for all subjects; the main difference was the value between 1 kHz and 7 kHz, where the impedance amplitude could be up to a factor 10 larger than that for a plane wave in air.

3.4. Clinical application

The amplitude of the specific acoustic impedance is compared with hearing thresholds in Fig. 7 for a subject with a hearing loss of 10 - 15 dB between 3 kHz and 10 kHz. A near perfect correlation is found between the two measures.

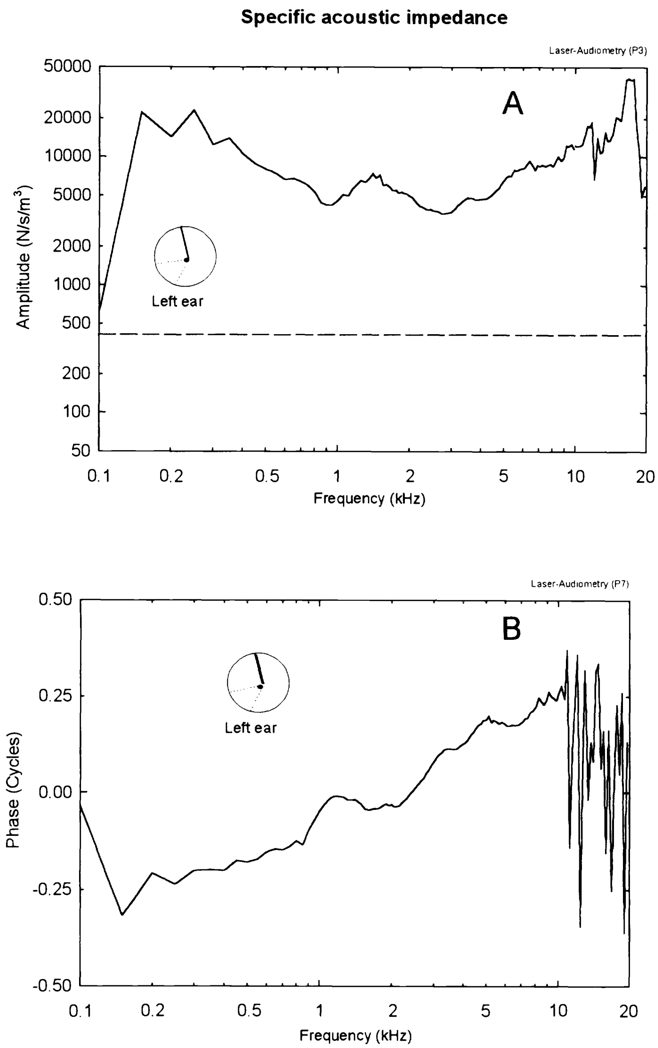


Fig. 6. Amplitude (A) and phase (B) of the specific acoustic impedance of the umbo. Dashed line: impedance for a plane wave in air.

4. DISCUSSION AND CONCLUSIONS

By coupling a commercially available laser Doppler velocimeter to a standard operations microscope, we have shown that it is possible to characterize middle-ear motion in human subjects non-invasively. The experiments show a strong correlation between the specific acoustic impedance of the umbo and hearing thresholds that may be employed for differential diagnosis.

The error due to the angle between the direction of the vibration and the direction of the laserbeam is minimal. For most measurements this angle was approximately 40° corresponding to an error of 2.5 dB.

In general, the vibration results are in agreement with those from other authors. The low-frequency value of 50 nm / 100 dB SPL was also found by Løkberg et al., (1980) [7] for humans *in vivo* using the holography method, for cats with laser interferometry [2,3,4], and for guinea pig with the Mössbauer method [6].

The different resonance peaks in the frequency response reflect the change in the vibration pattern of the umbo and eardrum with frequency and position. The first resonance frequency between 700 and 1200 Hz is in agreement with the results from other authors in temporal bone preparations [1] and *in vivo* [10]. This corresponds to middle-ear resonance as a result of tympanic membrane stiffness and mass.

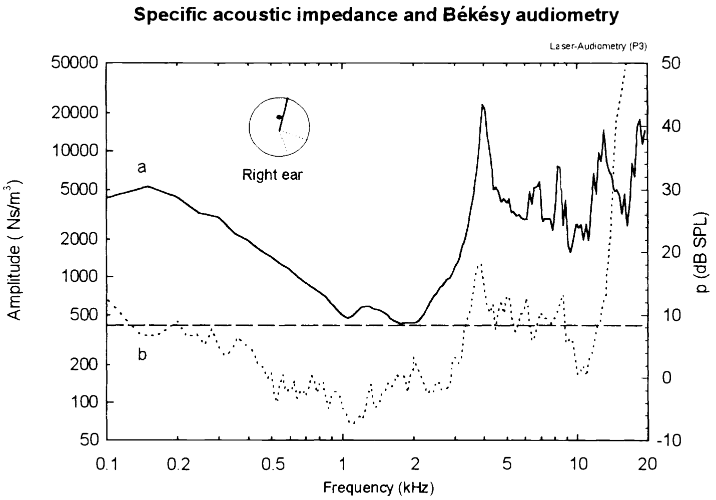


Fig. 7: Comparasion between (a) the specific acoustic impedance and (b) hearing thresholds for a subject with a hearing loss of 10-15 dB for 3-10 kHz

Løkberg (1980)[7] described a resonance in the region of 2 kHz, which he explained as being due to contractions of the muscles of the middle ear. This is possible because he stimulated with 90 dB SPL and the stapes contractions occur at 70 dB SPL above hearing threshold. However, although we stimulated at lower SPLs, we did find this resonance at 2 kHz in one subject.

The decline of the displacement amplitude of 6 dB/oct between 1 kHz and 7 kHz simply means that the specific acoustic impedance is constant in this frequency range. In other words, the ear appears to be adapted to speech frequencies. Importantly, the motion of the umbo is not only a function of middle-ear parameters but is also dependent on the input impedance of the inner ear. Changes in either middle- or inner-ear conditions should be reflected in the displacement response of the umbo. The strong correlation between specific acoustic impedance and hearing threshold (Fig. 7) is testimony to this fact.

Further experiments are being conducted to correlate middle-ear responses with middle- and inner-ear pathologies with the aim of using the described measurement system for differential diagnosis.

5. ACKNOWLEDGEMENTS

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