VIBRO-ACOUSTIC RESPONSE OF TYMPANIC-MEMBRANE-LIKE MODELS

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Equivalent measurements to the Real Ear Measurement (REM) procedure are sought using faster and contactless Laser Doppler Vibrometry. A feasibility study has been performed on tympanic-membrane-like rubber and latex models, in which the vibro-acoustic response was measured using both standard REM equipment and a PSV-500-HM Laser Doppler Vibrometer (LDV). The membrane surface velocity distribution was measured by the LDV and the transfer function between the forcing acoustic field and the vibration response of the membrane was calculated. The variability in the transfer function due to membrane shape, membrane type, and measurement position was determined. This study indicates that the transfer functions from measurements in the neighbourhood of the modelled umbo spot repeated several days apart were found to be consistent within typical clinical measurement tolerances. These results seem to indicate that Laser Doppler Vibrometry has the potential for replacing the standard REM procedure and equipment in patient-specific tests.

1. Introduction

The UK National Health Service (NHS) typically prescribes Behind The Ear (BTE) hearing aids that connect to a mould or an open fitting plug placed in the pinna, at the entrance of the ear canal, where the amplified sound is emitted and sent through the ear canal to the tympanic membrane [1 2]. In order to compensate the patient-specific hearing loss, a patient-specific Real Ear Measurement (REM) is performed by the audiologist for adult patients to calibrate hearing aids [1 3].

All data relating to normal hearing and normal ear acoustics are largely based on adults with little data available for children. This is because somatic, morphological and acoustical data are always changing for children due to their growth. The smaller ear canal of a young infant can result in a much higher Sound Pressure Level (SPL), up to +20 dB at 8 kHz [4], measured at the eardrum compared with an average adult’s attenuation and the variation with age and between children is larger than in adults, this being up to 35 dB of difference at certain frequencies [1 4].

To compensate for this change, the audiologist can either use a standard weighing function, created by averaging from a sample within the general population [1], or can perform a patient-specific Real Ear Measurement (REM) procedure. The option of using a standard weighing function in place of
REM is not available for children, owing to the rapid change in geometry of the ear canal that occurs in the first years of life. With adult patients, the REM procedure requires placing a probe microphone inside the ear canal, positioning it at a precise distance from the eardrum and sweeping the hearing aid output signal through the hearing frequency spectrum [2, 5]. This process takes few minutes, during which the patient is to remain still with the microphone in place. For this reason, this procedure is not generally performed on young children, as it would be impossible to obtain a reliable result unless the patient is sedated [1]. The current BTE hearing aid calibration in paediatric audiology is therefore typically performed without considering the influence of the ear canal and of its acoustic filtering effects, resulting in a sub-optimal calibration and hence degraded performance of the hearing aid [1].

The main objective of this study is to use the surface velocity of the eardrum as a proxy for the sound pressure level measured more intrusively by REM for specific subjects. This methodology can then be integrated within the standard procedures for individual fitting and calibration of hearing aid devices in paediatric and adult audiology [1]. This process assumes that a correlation can be established between the values of the SPL prescribed to a patient as verified by REM and the eardrum surface velocity. Therefore, to validate this hypothesis, preliminary experiments were carried out on models of the eardrum. Basic models were obtained by fitting latex and rubber sheets of three different thicknesses to hollow plastic tubes of three different shapes, representing different model ears. The vibrational velocity amplitude at the eardrum was then measured by Laser Doppler Vibrometry.

This feasibility study considers that a clinical audiologist will not be able to point the laser always at the same spot on the tympanic membrane of a patient in repeated measurements. It is therefore of interest to determine the variability in the measured velocity fluctuation amplitude from targeting different geometrical regions of the membrane. The thickness and shape of the tympanic membrane vary from one person to another and this feasibility study explores the effects of these factors on the measured vibrational velocity amplitude.

2. Methodology

The main challenge for this novel procedure is to estimate the acoustic impedance of the tympanic membrane by measuring the sound pressure acting on the membrane and the velocity at which it vibrates in response to the acoustic pressure. The tympanic membrane vibration amplitude is dependent on the level of sound pressure incident on the membrane. The acoustic characteristics of this pressure-velocity system can be first order approximated by a linear relationship through the transfer function $H(f)$, by which the velocity of the membrane $U(f)$ at frequency $f$ is related to the sound pressure $P(f)$ at frequency $f$ as $P(f) = H(f) U(f)$.

The transfer function for each model ear is derived by performing a set of experiments that measured the SPL in the hollow plastic tube, 5 mm lateral to the membrane, and the velocity of the model membrane synchronously. The ensemble averaged acoustic pressure amplitude from $n$ tests was evaluated as $\bar{P}^2(f) = \frac{1}{n} \sum_{i=1}^{n} P^2(f)_{i}$, where $P(f)_{i}$ is the acoustic pressure amplitude from the $i^{th}$ test in N·m$^{-2}$ at frequency $f$, in Hz. Similarly, the average vibration amplitude of the membrane from the same ensemble was evaluated as $\bar{U}^2(f) = \frac{1}{n} \sum_{i=1}^{n} U^2(f)_{i}$, where $U(f)_{i}$ is the vibration amplitude of the $i^{th}$ test in mm·s$^{-1}$. The transfer function for each model eardrum is then calculated as $H(f) = \bar{P}(f) \bar{U}(f)^{-1}$.

$H(f)$ provides the estimation of the acoustic pressure amplitude in each model ear, near the membrane, in a subsequent test, from just measuring the vibrational velocity amplitude of the membrane and using the relationship $P(f) = H(f) U(f)$.

To validate this approach, a second set of measurements was acquired of the velocity amplitude $U(f)$ and of the acoustic pressure amplitude $\bar{P}(f)$ near the model membranes, synchronously. The sound pressure amplitude near the membrane in each model ear was estimated as $\bar{P}(f) = H(f) U(f)$ and compared to the measured $P(f)$.

The difference $P(f) - \bar{P}(f)$ due to changes in the locality of the measurement (laser measurement spot), material thickness and shape are explored in Section 4.
3. Experimental setup

Figure 1(a) gives an overview of the experimental setup. The experiments were carried out in a quiet room to minimise background noise effects. Three hollow plastic tubes of length 5.0 cm and internal diameter of 1.5 cm were used to model the ear canal for each experiment. Each tube was installed as shown by arrow 4 in Fig. 1(a) and Fig. 1(b) gives a closer view of this installation. One of the tubes was left round at both ends, the second tube was flat-spotted into a D shape at one end and the third tube was flattened at one end. These plastic deformations resulted in the tube end profiles shown in Fig. 2(a). The out of round ends and one of the two round ends were covered with a membrane to model an eardrum. Thin 10.86 mg·cm$^{-2}$, benchmark 10.92 mg·cm$^{-2}$, and thick 28.78 mg·cm$^{-2}$ latex membranes were used in rotation on the end of the three tubes, as stated in Table 1, to investigate the effect of the material thickness on the vibration of the membranes. The membrane retainer did not include a system for adjusting the tension of each membrane and the latex membrane tension was controlled by not disturbing the installation between tests with the same membrane.

A clinical audiometer from AURICAL Plus, GN Otometrics, Denmark, shown by arrow 1 in Fig. 1(a), was used for producing an excitation sound signal external to the model ear canal and for measuring the sound pressure level inside the model ear canal close to the model tympanic membrane. White noise ranging from 40 dB to 80 dB and 100 Hz to 10000 Hz was generated by the audiometer through its speaker, shown by arrow 2, to excite the membrane. A microphone probe tube from REM equipment shown by arrow 4 measured the sound pressure level at about 5 mm lateral to the membrane in accordance with the Real Ear Measurement protocol. The device was interfaced to the desktop computer shown in Fig. 1(a) via the software application NOAH 4.0, HIMSA, Denmark. The instrumented model eardrum was clamped on the stand shown by arrow 6 and held with the open end at a distance of 0.5 m from the speaker with the tube axis perpendicular to the speaker cone axis.

A PSV-500-HM Laser Doppler Vibrometer from Polytec GmbH, Germany, measured the vibra-
Figure 2: Experiment 2 with benchmark circular membrane.

4. Effect of targeting

4.1 Effect of locality of targeting

Preliminary experiments aimed at qualifying the apparatus in Fig. 1 in terms of the repeatability of the measurements as well as the compatibility between the audiometer and the Laser Doppler Vibrometer instrumentations. These tests are referred to as Experiment 2 in Table 1 and use the round membrane made of benchmark material. This simple membrane geometry was selected as the vibrational behaviour of a thin circular membrane supported on a rigid ring is a classic problem in vibroacoustics that has an established analytic solution for the case of an isotropic elastic membrane [8].

Figure 2(b) shows the front-end view of the benchmark circular membrane mounted on the end of the 1.5 cm diameter round tube. The vibration of the membrane was surveyed by scanning over the pattern of points shown in Fig. 2(b). 12 scan target points create a circular pattern which increases radially outward, creating 9 concentric circles. These circles define the boundaries of concentric rings of 1.6 mm equal radial thickness. The central region, denoted by ‘C’ in Fig. 2(b), contains 25 scan points in a circle of 1.6 mm radius. Regions 1, 2 and 3 stand progressively from the centre outwards. The vibration measurements at the scan points approaching the membrane edge are not considered as they are significantly affected by the variability in the way the membrane is secured to the tube end.

Figure 3(a) shows the SPL measured by the microphone tube of the REM equipment from three
tests performed on different days. The measurements show that the REM equipment was able to provide a consistent level of excitation to the membrane among the different tests and gave confidence that variations in the measured vibrational velocity amplitude can be attributed to the mechanical response of the membrane as opposed to changes to the input sound field. The sound pressure level spectra in Fig. 3(a) peak at 1 kHz and 2 kHz. These peaks are consistent with the frequency range for the onset of longitudinal resonances in a 5 cm long rigidly-terminated tube, which would be approximately 1.7 kHz for the quarter-wavelength mode at ambient conditions. Higher order modes were observed on the membrane from 2.2 kHz. At this frequency, two out of plane velocity maxima located symmetrically either side of the centre suggest the presence of the second spinning mode.

Figure 3 shows the response of the circular membrane to the incident sound field of Fig. 3(a). Measurements from points belonging to the same region have been ensemble averaged to obtain the mean vibrational velocity spectrum at increasing radial distance from the centre of the circular membrane. All spectra appear to document fundamentally the same kind of mechanical response of the membrane, in that the spectra display five broad peaks over the frequency range 400 Hz to 6 kHz. The vibrational amplitude at these frequencies progressively decreases from the centre region C to region 3. The largest amplitude measurement from the centre region C was selected as the reference response $U(f)$ for determining the transfer function $H(f)$ according to the procedure of Section 2.

Estimates of the SPL at the eardrum were then computed by repeating the same measurements and using the estimator $\hat{P}(f) = H(f)U(f)$, with $U(f)$ measured in the central region and in the peripheral regions. $\hat{P}(f)$ displayed some variability depending on the measurement zone of $U(f)$.

(a) Sound Pressure Level (SPL) measured by the REM probe in three similar experiments and their average. (b) Velocity measured at different regions for circular tube.

Figure 3: Measurements from experiment 2 with benchmark circular membrane.

(a) Variability of SPL estimation from different regions with respect to the central region SPL estimate. Benchmark membrane model. (b) Variability of SPL estimation from different regions with respect to the central region SPL estimate. Thin membrane model.

Figure 4: Experiments with circular membrane.
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Figure 4(a) analyses this difference in detail. Specifically, Fig. 4(a) presents the difference in dB between the SPL estimated from vibrational velocity amplitude measurements at the central region and that from progressively more peripheral regions over the frequency range most relevant to human speech recognition. The difference among the zones is limited to ±1 dB within the typical hearing aid range of 500 Hz to 6 kHz. Greater discrepancies of up to ±3.2 dB are shown below this range. This is very satisfactory and indicates that the measured vibrational response of the benchmark round membrane is rather consistent within the scanned area. This allows inferring that the assessment of the vibrational response of the model eardrum is relatively insensitive to the location of the measurement within reasonable bounds, so that a protocol for measuring the response of a human tympanic membrane by LDV may include some flexibility in how accurately the LDV has to aim at its target.

4.2 Material and shape effects on targeting

4.2.1 Circular membrane of thin material

Somatic variations between individuals determine an appreciable variability in the eardrum thickness among the human population. This affects the membrane stiffness. Experiment 1 in Table 1 considers whether a broad targeting LDV procedure can be applied to young patients, who would typically have thinner eardrums. The vibrational response of a 10.86 mg·cm⁻² thinner latex membrane was assessed by measuring its out of plane velocity using the same scan pattern shown in Fig. 2(b). A transfer function \( H(f) \) was obtained from the central region data that is specific to the thinner latex membrane using the procedure of Section 2. Different estimates \( \hat{P}(f) \) of the SPL at the eardrum where then obtained from subsequent LDV measurements from regions C, 1, 2, and 3. The difference in the estimated SPL spectra obtained from ensemble averaging the measurements in the central region and in the more peripheral regions is shown in Fig. 4(b). As in Fig. 4(a) the differences are stated in dB with respect to the SPL spectrum estimated from the central region \( U(f) \). Figure 4(b) indicates that the thinner membrane gives more variability in the vibrational response with changes in the LDV spot location. The significance of this difference between the thin and benchmark membrane responses was examined by determining the p-value of the differences in each region, against the t-distribution \( p_{0.05} = 1.81 \), over the range 500 Hz to 6 kHz. In regions 1, \( p = 0.03 \) and the difference is not significant. The more off-centre LDV measurements from regions 2 and 3 give significant differences, with \( p = -2.5 \) and \( p = -6.5 \) respectively, reflecting an up to 7 dB difference in region 3. This result is less satisfactory but can still be considered an improvement from the current practice of not performing REM measurements in paediatric audiology or applying ear canal attenuation corrections derived from adult data.

4.2.2 D-shaped membrane of thin material

Having established that consistent predictions of SPL can be obtained, within bounds, from the vibrational characteristics of a membrane of simple geometry, it is of interest to test whether the technique can be applied to more irregularly shaped membranes. The variability in the predicted SPL from scanning a D-shaped membrane was determined in Experiment 4 of Table 1. The Cartesian scan grid shown in Fig. 5(a) was used and one central and one peripheral zone were defined around the area of largest vibration amplitude, shown in red in Fig. 5(a). The outer region is 1.5 mm away from the centre. Estimates of the SPL at the eardrum are obtained using \( H(f) \) from the central region measurements and the effect of using off target \( U(f) \) measurements from region 1 is assessed in Fig. 5(b). With the D-shaped membrane, using surface velocity measurements from the peripheral region 1 for estimating the SPL at the eardrum results in an error of ±1.5 dB across the frequency range 100 Hz to 10 kHz, which includes the typical hearing aid frequency calibration range of 500 Hz to 6 kHz. This very satisfactory result indicates that the more irregular shape of the membrane in Fig. 5(a) has decreased the sensitivity of the measurement procedure on the locality of the targeting.
This is possibly related to the suppression of the circular membrane natural resonances by the shape change, which is possibly the same process observed in nature by which human tympanic membranes have evolved into bean shaped membranes.

4.2.3 Flattened tube with benchmark membrane

The positive effect that an out of round membrane shape has on decreasing the sensitivity of the SPL estimate on the LDV targeting accuracy prompted a further experiment to determine whether this positive outcome is shape-specific. In experiment 5 of Table 1, a membrane of benchmark material was applied to a flattened tube, resulting in the irregular membrane outer profile of Fig. 2(a). The profile is approximately a barrel shape with two horizontal flattened edges. A Cartesian scan pattern is defined on the membrane to include the majority of the mobile surface. Velocity data from the LDV verified that all the areas of maximum amplitude, shown in red in Fig. 6(a), are contained within the scan area. As in experiment 4, a central region C and a peripheral region 1 are defined. The peripheral region extends 1.25 mm away from the central region. Measurements of surface velocity were obtained in these regions from which estimates of the SPL near the model eardrum were obtained following the procedure of Section 2. Figure 6(b) shows that the variations in SPL estimated from velocity measurement inputs performed on the periphery, in region 1, are limited to ±1 dB across the typical hearing aid calibration frequency range of 500 Hz to 6 kHz. Outside this frequency range, the variability increases to ±2.2 dB. This experiment suggests that the lower sensitivity of the predicted SPL on the LDV target location is a relatively general feature of irregularly shaped membranes. This is a positive outcome, in view of the human tympanic membrane being bean shaped.

5. Conclusions

This study has completed the first step towards solving the current limitation in paediatric audiology of not being able to perform routine REM measurements on the young patients. The feasibility of a contactless and non-invasive alternative procedure to REM has been investigated using basic model eardrums. It was found that, within bounds, the vibrational response from the model eardrum could be used as a pseudo variable for estimating the SPL near the model tympanic membrane. It was observed that the SPL predictions had good repeatability even if the measurement spot on the membrane had some variability, anticipating what can realistically be achieved with young patients in paediatric practice. Variations of approximately ±1 dB were found within the typical hearing aid range working frequency of 500 Hz to 6 kHz in tests involving irregularly shaped model membranes.
Whilst these preliminary results are encouraging, the basic ear model in these tests did not account for the effect of the air pressure in the middle ear, the anisotropic thickness of the human membrane, the presence of pathologies of the middle ear ossicles, and other factors that may influence the development of an effective and clinically robust measurement protocol. Therefore, further work is required for validating the use of LDV as a REM proxy for fitting better calibrated hearing aids in paediatric audiology.

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REFERENCES