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Investigation of a Novel Completely-In-The-Canal Direct-Drive Hearing Device: A Temporal Bone Study

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Abstract

Hypothesis—Whether a prototype direct-drive hearing device (DHD) is effective in driving the tympanic membrane (TM) in a temporal bone specimen to enable it to potentially treat moderate to severe hearing loss.

Background—Patient satisfaction with air conduction hearing aids has been low due to sound distortion, occlusion effect, and feedback issues. Implantable hearing aids provide a higher quality sound, but require surgery for placement. The DHD was designed to combine the ability of driving the ossicular chain with placement in the external auditory canal.

Methods—DHD is a 3.5 mm wide device that could fit entirely into the bony ear canal and directly drive the TM rather than use a speaker. A cadaveric temporal bone was prepared. The device developed in our laboratory was coupled to the external surface of the TM and against the malleus. Frequency sweeps between 300 Hz to 12 kHz were performed in two different coupling methods at 104 and 120 dB, and the DHD was driven with various levels of current. Displacements of the posterior crus of the stapes were measured using a Laser Doppler Vibrometer.

Results—The DHD showed a linear frequency response from 300Hz to 12kHz. Placement against the malleus showed higher amplitudes and lower power requirements than when the device was placed on the TM.

Conclusions—DHD is a small completely-in-the-canal device that mechanically drives the TM. This novel device has a frequency output wider than most air conduction devices. Findings of the current study demonstrated that the DHD had the potential of being incorporated into a hearing aid in the future.

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INTRODUCTION

Hearing loss is one of the most prevalent chronic conditions in the U.S., affecting over 16% of the population [1, 2]. The decision to provide hearing impaired patients with acoustic amplification depends on the age, the degree of hearing loss, and the individuals' self perceived communication difficulty [3]. While hearing aids are available to assist sufferers of hearing loss, fewer than 20% of people who needed them actually owned a hearing aid in 2004 [4]. Despite technological improvements in conventional hearing aids and the recent introduction of middle ear implants, the obstacles to hearing aid adoption remain.

Patient satisfaction with air conduction hearing aids has remained low due to lifestyle restrictions, sound distortion, occlusion effect, and feedback issues [5, 6]. A common solution for the occlusion effect is to add a vent to the ear mold to allow the sounds trapped in the ear canal to escape [7]. The vent, however, reduces the attenuation from the speaker to the microphone and increases the likelihood of feedback [8]. The feedback is overcome by increasing the separation between the microphone and speaker, usually by increasing the size of the hearing aid (in order of increasing size and visibility) [6]. However, patients generally do not desire to wear larger hearing aids due to their appearance and attached stigma. Although digital feedback management techniques can be applied, the state-of-the-art feedback management algorithms have been reported to lead to signal degradation [7].

Small hearing aids have been developed, but they suffer from the feedback problem just described. One of the newer hearing aids on the market (Lyric, InSound Medical Inc., Newark, CA) is small enough to be inserted deep into the bony part of the ear canal without being visible. The device eliminates the stigma attached to hearing aids and reduces the occlusion effect by reducing the amount of sound generated in the ear canal. However, due to potential feedback problems associated with a short distance between microphone and speaker, the device is only indicated for people with mild to moderately severe hearing loss who do not require significant amplification [9].

Implantable hearing aids, on the other hand, have provided a higher quality sound, and overcome the aforementioned drawbacks. Undesirably, middle ear implants suffer from prohibitive cost, the need for surgery (and potential complications), and almost all of them have been designed for patients with moderate to severe sensorineural hearing loss with discrimination scores of more than 40–60%. [10, 11].

We have attempted to address the above-mentioned drawbacks of the commercially available hearing aids by combining the advantages of various types of hearing aids into one model. In our laboratory, we have been designing and developing a novel direct-drive hearing device (DHD) that recreates sound with mechanical movements of the tympanic membrane (TM). DHD operates similar to a middle ear implant, except that it is placed on the external surface of the TM and, therefore, does not require surgery. Placement of the device within the bony portion of the ear canal and stimulation of the malleus using a mechanical actuator could minimize the occlusion effect. The direct mechanical drive could eliminate acoustic feedback problems, and allow improved sound quality and increased functional gain. In the current study, we aimed to test the DHD on a cadaveric temporal bone to evaluate the effectiveness of the mechanical actuator in driving the TM and its optimal positioning to prepare the device for pre-clinical experiments.

METHODS

Direct-Derive Hearing Device Description

The DHD was 6.2 mm long and had a 3.7 mm diameter that enabled it to entirely fit into the deep osseous external ear canal (Figure 1). A schematic drawing of the eventual iteration of the device is depicted in Figure 2. There are four major components: (1) TM/malleus driver, (2) microphone, (3) signal processor and (4) power system. The most significant and distinguishing component that was tested in the current study was the malleus driver, which is a voice coil actuator. The malleus driver was composed of a permanent magnet, two flux components to guide the magnetic field, a voice coil, and a contact tip. This design created a motive force that drove the TM through the contact tip by the reaction of the magnetic field of the permanent magnet and the current carrying voice coil. The engineering design, implementation and validation of the system and its individual components will be published elsewhere.

Outcome Measurements

The device was validated to ensure if, first, it was capable of providing mechanical movement across various frequencies, and second, the movements will adequately recreate sounds without redundant noise.

The frequency response was measured from 300 Hz to 20 kHz in 1/6 octave steps using laser doppler vibrometer (LDV) system (MSA-500 Micro System Analyzer, Polytec; Irvine, CA). The LDV uses a precise laser wave to monitor the Doppler shift of a moving object. A small piece of reflective tape (Polytec; Irvine, CA) was placed on the contact tip. Then, the device was placed under the LDV and laser was shined on the reflective tape to record the movements.

The acoustic noise generation of the device was recorded by a calibrated ER-7C probe microphone (Etymotic Research, Inc.; Elk Grove Village, IL) at 1 mm distance and analyzed using SoundCheck 8.0 (Listen, Inc.; Boston, MA). The effectiveness of the DHD in reproducing sounds was determined by quantification of its ability to induce ossicular chain vibration of a human temporal bone similar to a previously described method [12, 13].

Temporal Bone Preparation

Upon obtaining the Institutional Review Board approval, a left cadaveric temporal bone of a 59-year-old male with no history of middle ear diseases (eight years post mortem) was obtained from the Willed Body Program at our institution. The bone had been fixed in a 10% solution of neutral buffered formalin and was kept at a temperature of 4 °C. After securing the temporal bone in a bone holder, a simple mastoidectomy with facial recess approach was performed. The posterior crus was selected to measure the velocity. A small piece of reflective tape was placed on the surface of the posterior crus of the stapes for subsequent LDV measurements. A cosine correction of the measured displacement was applied to account for the measurement angle offset. An offset angle of 45° was determined from the average of two independent observers with a 5° difference.

Ossicular Displacement Measurement

The temporal bone was positioned and fixed under the LDV to enable illumination of the laser beam on the posterior crus (Figure 3). Then, the LDV was used to measure the displacements of the stapes caused by vibration of the TM under 4 different settings. First, the displacements caused by the background noise were measured to account for their potential effect on the measurements in the subsequent settings. Second, baseline measurements were acquired to simulate natural hearing using a calibrated ER-5A insert

earphone (Entymotic Research, Inc.; Elk Grove Village, IL) placed at 2 mm distance from the TM. Pure tone sound stimuli were delivered from 200 Hz to 12 kHz in 1/6 octave steps at 104 and 120 dB SPL (measures simultaneously with an ER-7C probe microphone) and the displacements were recorded. The baseline measurements allowed the middle ear's responses to acoustic stimulation and the DHD's mechanical stimulation to be compared. Also, the baseline measurements allowed the effects of size, age, and other physiological variability to be taken into account.

Third, the DHD was carefully inserted inside the osseous external auditory canal, the contact tip was placed on the posterior-superior quadrant of the TM under direct microscope, and the device was secured using bone wax. The DHD was stimulated with a stepped sine wave from 300 Hz to 12 kHz in 1/6 octave steps and stimulating input at 200, 400, 600, 800 mV voltage levels. Fourth, the same procedure was repeated except that the DHD contact tip was placed on the umbo.

Statistical Analysis

The mean, standard deviation, and range of displacements were calculated for each hypothetical setting. Because of non-normal distribution of the displacements, Mann-Whitney U test was implemented to compare the displacements between the settings in which the DHD was coupled to the TM and to the umbo. Kruskal-Wallis H test was implemented to compare the displacements between background noise and different voltage inputs of the device within each model of coupling. All statistical procedures were performed using PASW 18.0 (SPSS Inc., Chicago, IL). A *p* value of less than 0.05 was considered to be statistically significant.

RESULTS

Validation of the DHD prior to cadaver tests revealed that the displacements of the contact tip and the frequency had a negative linear response from 300 Hz to 20 kHz (range: 0.9 to 9,000 nm). Total harmonic distortion of device was less than 0.5% above 400 Hz. The maximum distortion was 2.3% at 300 Hz. Acoustic noise generation of the device was measured in a quiet room and the device did not produce noise more than the background noise, which was between 30 and 35 dB across all frequencies (see Video, Supplemental Digital Content 1, which demonstrates the capability of the actuator in playing music through a soft surface that simulates the TM).

Figure 4 and 5 demonstrate the results of the displacements of the posterior crus measured by the LDV when the DHD was coupled to the posterior-superior quadrant of the TM and the umbo respectively compared to background noise and baseline measurements at 104 and 120 dB SPL. Table 1 exhibits the mean, standard deviation, and range of displacements in different settings. The range of displacements of the stapes with hearing simulation using sound pressure of 104 dB SPL was 0.34 – 10.02. This range was most comparable to the 400 mV input of the DHD where it was coupled to the TM (0.11 – 24.11), and to the 200 mV input of the DHD where it was coupled to the umbo (0.38 – 26.62). Similarly, the range of displacements in 120 dB SPL simulation (2.56 – 63.48) was most comparable to the 800 mV input of the DHD where it was coupled to the TM (0.33 – 54.45), and to the 400 mV input of the DHD where it was coupled to the umbo (0.64 – 55.22). All differences between and within various settings were significant ($p < 0.001$).

DISCUSSION

The direct-drive hearing device, DHD, is a novel, unconventional and innovative device that has been designed to address the issues associated with feedback, complications, and

visibility of the commercially available hearing devices. Validation of the DHD demonstrated a linear frequency response and total harmonic distortion of less than 0.5% above 400 Hz, corresponding to the previous recommendations for implantable hearing aids [14]. The higher distortion below 400 Hz could have been a result of the intrinsic characteristics of the device or the change in the length of the laser used by LDV (20 $\mu\text{m}/\text{V}$ for frequencies below 3000 Hz and 0.5 $\mu\text{m}/\text{V}$ for frequencies above). There was a wide gap between the amount of noise generated by the device (<35 dB SPL) and the induced displacements of the posterior crus of the stapes (equivalent to baseline 104 dB SPL). The linear frequency response, low total harmonic distortion, and small acoustic noise production of the DHD makes it a competitive alternative to air conduction or implantable hearing aids. These findings compared well with those of the Vibrant Soundbridge device tested in our laboratory (data not presented). Park et al. [15] also tested a novel differential electromagnetic floating-mass transducer and found it to have a negative linear frequency response above 1 kHz. The stapes displacement in the DHD ranged between 1 and 200 nm, and was comparable to those of the previous reports. Javel et al. [16] tested two piezoelectric transducers with an input of 1 V. The displacements were measured at the malleus level of an anesthetized cat and ranged between 1 and 100 nm. In another study, Huber et al. [17] assembled a middle ear prosthesis from a Vibrant Soundbridge and a Bell Tübingen titanium prosthesis, and tested it on a fresh cadaver. Their prosthesis was stimulated with 100 mV and was able to induce displacement of the stapes footplate in fresh cadaver bone, which ranged between 1 and 300 nm.

The DHD mechanically drove the ossicular chain at the TM level. An early attempt on driving the umbo placed a small magnet driven by an external coil on the TM by surface adhesion, using mineral oil as a surface wetting agent [18]. One problem with the earlier device was that the amplitude of the magnet's vibration varied with the position of the coil, and quickly decreased with increasing distance, thus increasing the power requirements of the device. A second problem was that the magnet attached to the TM loaded the ossicular chain, reducing the stapes displacement especially at higher frequencies [19]. Mass loading as little as 25 mg can cause a reduction in stapes displacement [20]. This resulted in even more hearing loss when the device was turned off.

Currently, there is no available practical method for attaching a magnet to the TM for long period. In our device, both the coil and the magnet were housed within the same assembly. This allowed us to optimize both space and geometry, and provided a controllable and stable vibration. Furthermore, with the coil wound around the magnet and related flux circuit, the required current needed to drive the magnet was small. The DHD will not cause a mass loading; therefore, the ear will be unaffected by the device when it is not turned on. Our results demonstrated that coupling the contact tip of the device to the umbo required half the voltage to deliver the same energy as compared to coupling to the TM itself.

The safety and feasibility of umbo vibration have been reported in a clinical study on a malleus vibration audiometer (MVA) as a diagnostic tool for assessing ossicle function and integrity [21]. The MVA presented mechanical vibrations to the umbo via a stiff coupling rod about the length of the ear canal with a specially shaped tip. The study found no pure-tone threshold shifts after 25-minute measurement procedures performed on two days. Three out of 38 subjects reported slight pain, but not more than the pain experienced during a standard ear exam with an ear speculum. No other adverse events occurred. None of the subjects reported any ear-related symptoms and no injuries on the TM were found.

It has been found that measures of middle-ear input (i.e., umbo velocity) are approximately similar between live and fresh human cadaver ears [22, 23]. Employing a fixed cadaveric temporal bone for testing the DHD may limit the results of the current study since

theoretically, a fixed middle ear may result in lower magnitudes of displacements. Nonetheless, this would not invalidate the comparisons of different studied models since the comparisons were made to air conduction sound in the same cadaver. The pattern of posterior crus displacements by the DHD differed from the baseline measurements by sound pressure. Since the uncoupled device showed a linear response prior to coupling to the TM, it is plausible that the dynamics of the interaction between the contact tip and the TM have caused this difference. However, the pattern of displacements was almost similar when the device was coupled to the posterior superior quadrant compared to when it was coupled to the umbo.

Future development of the DHD needs to consider the challenges associated with the long-term use of completely-in-the-canal (CIC) hearing devices, as indwelling foreign objects in the ear canal. Pain, irritation, keratin and moisture build up may occur [6]. In addition, various shapes and curvature of the bony canal may limit deep insertion of the devices [24]. It has been reported that 18% of the Lyric hearing aid users reported pain in the ear, irritation in the ear canal, and occlusion or a build-up of moisture between the hearing aid and the eardrum [25]. However, no signs of inflammation, eardrum perforation or sudden deterioration in the user's ability to hear were noted, and all of the symptoms were relieved within five days. Most of these users were found to have a narrow diameter or a curvature in their ear canal. Long-term studies are needed to evaluate the effects of direct contact of this device with the TM before the device can reach wide clinical use.

Given the characteristics of the DHD, its application may not be limited to acoustic amplification. The malleus driver could also attenuate the passage of acoustic energy onto the ossicles and work as an active earplug. Future studies of the DHD will focus on the dynamics of the device-TM interface and fixing the device inside the ear canal. The dynamics of the device-TM interface will be investigated through testing several fresh cadaveric temporal bones to define the average range of frequency responses and stapes displacements. Furthermore, different forms of the contact tip will be tested to optimize the interface.

CONCLUSION

This was a temporal bone study that evaluated a novel, small, and invisible device with potential of being incorporated in a hearing aid in the future. The DHD had linear frequency response, and low noise generation and total harmonic distortion. The device could be placed within the osseous external ear canal of the temporal bone in contact to the TM to directly drive the TM. The mechanical actuator was effective in transmitting the vibrations onto the ossicles at a similar level as an implantable hearing aid. Coupling the DHD to the umbo instead of the peripheral surface of the TM resulted in less energy consumption.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1.
The pioneer direct-drive hearing device that was tested on the cadaver temporal bone compared to a paper clip.

Figure 2.
Schematic design of the direct-drive hearing device.

Figure 3.
The temporal bone placed under direct illumination of the laser Doppler vibrometer. The reflective tape on the surface of the posterior crus is shown on the monitor.

Figure 4.

Frequency response of the direct-drive hearing device at different voltage inputs when the device was coupled to the posterior-superior quadrant of the TM, compared to the background noise and the baseline measurements at 104 and 120 dB SPL.

Figure 5.

Frequency response of the direct-drive hearing device at different voltage inputs when the device was coupled to the umbo, compared to the background noise and the baseline measurements at 104 and 120 dB SPL.

Table 1

Mean, standard deviation and range of displacements the posterior crus of stapes across different settings of the study.

	Device coupled to the tympanic membrane (nm)		Device coupled to the umbo (nm)		<i>p</i> value
	Mean $\pm$ Standard Deviation	Range	Mean $\pm$ Standard Deviation	Range	
Background	0.06 $\pm$ 0.11	0.00 – 0.59	0.06 $\pm$ 0.11	0.00 – 0.59	-
104 dB SPL	3.19 $\pm$ 2.83	0.34 – 10.02	3.19 $\pm$ 2.83	0.34 – 10.02	-
120 dB SPL	23.83 $\pm$ 18.40	2.56 – 63.48	23.83 $\pm$ 18.40	2.56 – 63.48	-
200 mV	0.88 $\pm$ 1.08	0.02 – 3.32	8.17 $\pm$ 5.84	0.38 – 26.62	<0.001
400 mV	5.62 $\pm$ 7.10	0.11 – 24.11	23.26 $\pm$ 17.34	0.64 – 55.22	<0.001
600 mV	7.98 $\pm$ 8.68	0.20 – 29.08	35.88 $\pm$ 27.82	0.89 – 82.54	<0.001
800 mV	13.03 $\pm$ 15.66	0.33 – 54.45	53.76 $\pm$ 48.77	1.11 – 197.20	<0.001
<i>p</i> value *	<0.001	-	<0.001	-	-

* These *p* values demonstrate the statistical difference between the background noise different voltage inputs. Baseline measurements were not included.