厚生労働科学研究研究費補助金

感覚器障害研究事業

軽量コイルにより耳小骨を直接加振する
新駆動方式 Hi-Fi 補聴システムの開発に関する研究

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I. 総括研究報告書

軽量コイルにより耳小骨を直接加振する
新駆動方式 Hi-Fi 補聴システムの開発に関する研究

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Abstract

Research and development over the past two decades has shown that implantable hearing aids can circumvent some of the problems found in conventional hearing aids, such as feedback, cosmetic problems and so on. However, these hearing aids have not as yet been widely used, because invasive surgery has to be performed to implant them and applying them in children, who are still growing, is difficult.

In this study, a prototype of the non-implantable electromagnetic hearing aid, which can generate a high-excitation force to vibrate the ossicles via the tympanic membrane, was constructed. The hearing aid consists of an electromagnetic transducer and an amplifier. To determine fundamental properties of the hearing aid, excitation force and acoustical gain were evaluated using an artificial middle ear and a human temporal bone, respectively. The experiments showed that the hearing aid was able to generate the maximum excitation force, which is equivalent to a sound pressure of 100 dB SPL at high frequencies.

Chapter 1. Introduction

Recently, sensorineural hearing loss, resulting from the loss of the function of cochlear amplification, is observed in a steadily increasing number of younger patients as well as in aged patients [12]. The majority of these hearing-impaired individuals use conventional hearing aids, which use earphones. However, part of patients who use conventional hearing aids are not satisfied yet. The dissatisfaction with using conventional hearing aids are owing to many inherent problems such as sound distortion, feedback, cosmetic factors and so on.

Research and development over the past two decades has shown that implantable hearing aids can circumvent some of the problems found in conventional hearing aids. The most prominent feature of the implantable hearing aid is that an actuator is directly coupled to the one of the middle-ear ossicles. It has the advantage of leaving the ear canal open, and problems with feedback can be eliminated.

There are two major methods for achieving excitation, namely, by way of piezoelectric transducers and electromagnetic transducers. Suzuki and Yanagihara et al. [7, 13] developed a piezoelectric transducer. The transducer is a piezoelectric ceramic bimorph attached to the stapes to vibrate it. Electromagnetic hearing aids consist mostly of a permanent magnet placed on the ossicles and an inductive coil to drive the magnet. Maniglia et al. [10] reported an electromagnetic transducer which employs a magnet attached to the body of the incus. Moreover, Vibrant Soundbridge® is commercially available now [9]. It uses a transducer named Floating Mass Transducer™, which is a magnet surrounded by two induction coils in a titanium container. The transducer is attached to the long process of the incus by a clip, and the vibrations of the magnet are transmitted to the ossicles.

Besides the above-mentioned devices, many types of piezoelectric or electromagnetic transducers have been developed [2, 3, 4, 5, 6, 11, 14, 15, 16]. However, implantable transducers have not as yet been widely used, because invasive surgery has to be performed to implant them. Therefore, high-fidelity hearing aids, which are non-implantable, are needed.

In this study, a non-implantable electromagnetic hearing aid, which can generate a high-excitation force to vibrate the ossicles via the tympanic membrane, was constructed. The hearing aid consists of an electromagnetic transducer and an amplifier. To determine fundamental properties of the hearing aid, an excitation force generated by the hearing aid was evaluated using (i) artificial middle ear and (ii) human temporal bone.

Chapter 2. Non-implantable electromagnetic hearing aid

The non-implantable electromagnetic hearing aid developed in this study is shown in Fig. 2.1(a). It can be divided into two parts, i.e., an electromagnetic transducer and an amplifier. The transducer was designed so that it can be installed into the human external ear canal.

Figure 2.1(b) shows the principle of the transducer. The transducer is composed of a core, driving and induction coils, a rare earth magnet, and a vibrator coil. The induction coil is placed in the external ear canal. The magnet is fixed by a holder near the tympanic membrane. The vibrator coil adheres to the tympanic membrane by means of oil. When an alternating current is supplied to the driving coil, magnetic flux passes through the core. The induced electromotive force is then generated in the induction coil. The induced current that flows into the vibrator coil causes magnetic interaction between the magnetic field generated by the vibrator coil and the static magnetic flux of the magnet. As a result, the vibrator coil, which is attached to the tympanic membrane, is excited by the repulsive and attractive forces, which act between the magnet and the vibrator coil.

Figure 2.2 shows the photograph of the hearing aid. The amplifier in this hearing aid is an existing amplifier (RION Co. Ltd.) which is used in other middle ear implant. Co-based amorphous magnetic material with high permeability was used for the core material and the driving coil was wound around it. Its diameter and length are 1.5 mm and 30 mm, respectively. A cylindrical shaped Ne-Fe-B magnet with a surface magnetic density of 1.2 T is used in this study. The diameter and thickness of the magnet are 3 mm and 1.5 mm, respectively. The driving coil contains 2000 turns of insulated copper wire with a diameter of 60 μm. Its length is the same as that of the core. The induction coil contains 100 turns of insulated copper wire with a diameter of 140 μm, and its inner and outer diameters are 4 mm and 6 mm, respectively. Its length is 5 mm. With regard to the vibrator coil, it contains 14 turns of insulated copper wire with a diameter of 60 μm. The vibrator coil was attached to a plastic plate, which is used as the attachment to the tympanic membrane. Total mass of the vibrator coil and plastic plate was 18.3 mg.

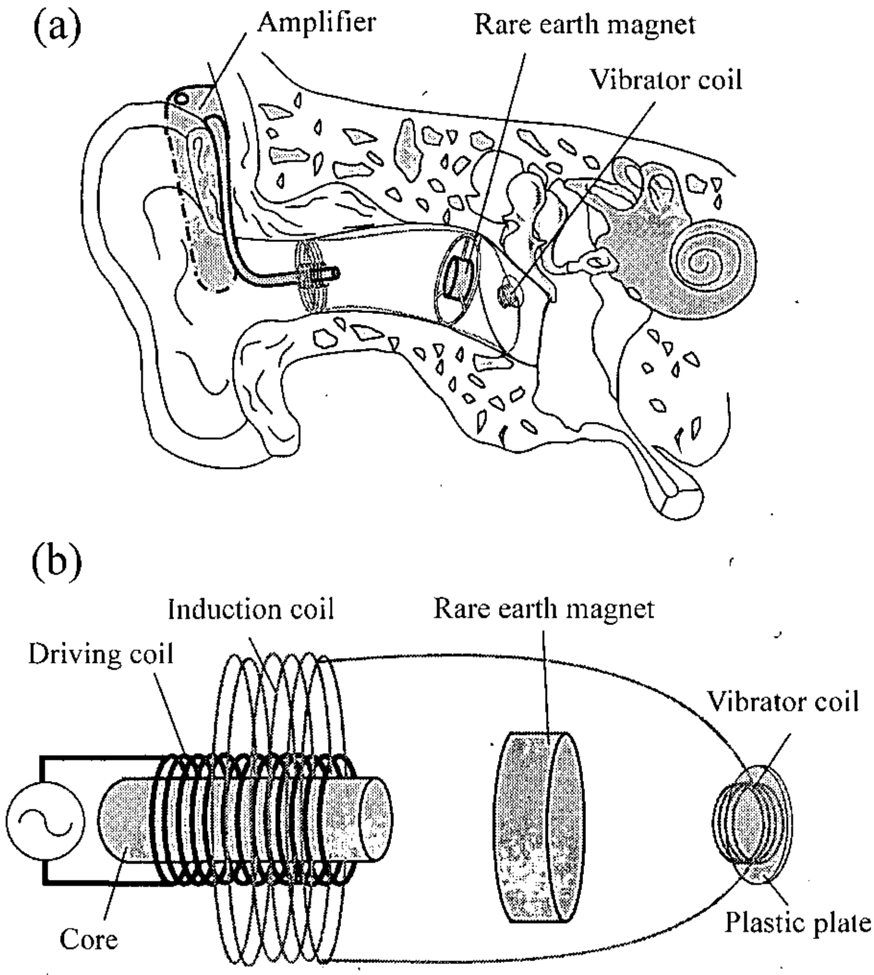

Figure 2.1. Non-implantable electromagnetic hearing aid. (a) Schematic of the non-implantable hearing aid developed in this study. The amplifier, the core and the driving coil are easy to wear and remove. (b) Principle of the electromagnetic transducer. An alternating current generated in the induction coil flows into the vibrator coil causes the repulsive and attractive forces, which act between the magnet and the vibrator coil. The driving and induction coils and the magnet were designed so as to generate the greatest excitation force.

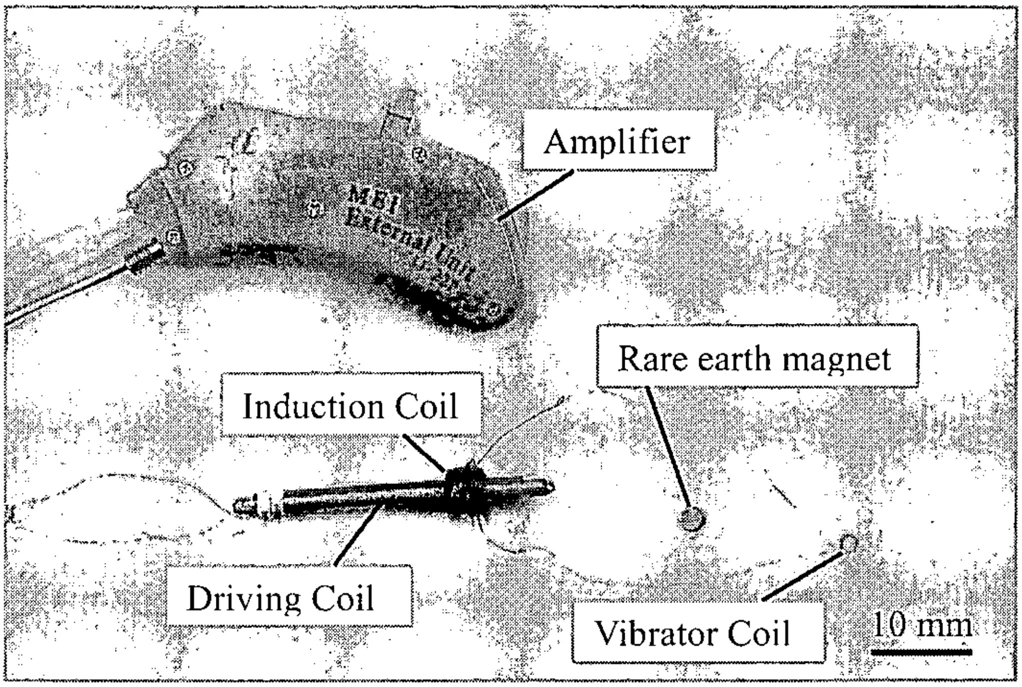

Figure 2.2. Photograph of a prototype of the hearing aid developed in this study. With regard to the amplifier, an existing amplifier (RION Co. Ltd.), which is used in other middle ear implant was adopted. Co-based amorphous magnetic material with high permeability was used for the core material and the driving coil was wound around it. A cylindrical shaped Ne-Fe-B magnet with a surface magnetic density of 1.2 T is used in this study. The diameter and thickness of the magnet are 3 mm and 1.5 mm, respectively. The vibrator coil contains 14 turns of insulated copper wire with a diameter of 60 μm. The vibrator coil was attached to a plastic plate, which is used as the attachment to the tympanic membrane. Total mass of the vibrator coil and plastic plate was 18.3 mg.

Chapter 3. Methods

3.1. Artificial middle ear

An attempt was made to evaluate the fundamental properties of the hearing aid using an artificial middle ear. The artificial middle ear, shown in Fig. 3.1(a), is composed of a plastic tube (8 mm in diameter and 15 mm in length), a silicone membrane (80 μm in thickness) and a plastic chip made of polypropylene, which corresponded to the external ear canal, the tympanic membrane and the malleus, respectively. The total mass of the membrane and chip was 42 mg, the same as the mass of the human tympanic membrane and ossicles.

To evaluate the excitation force, sound pressure of 60 dB SPL was applied to the microphone of the hearing aid by an earphone (ER-10C, ETYMOTIC RESEARCH), and frequency responses of the displacement at the tip of the plastic chip of the artificial middle ear were measured with a laser Doppler velocimeter (LV-1400, ONO SOKKI). In addition, the hearing aid was removed and the earphone was inserted into the plastic tube and the displacement was measured when a constant sound pressure of 80 dB SPL was applied to the artificial middle ear by the earphone, as shown in Fig. 3.1(b).

3.2. Human temporal bone

Although the estimation of the excitation force generated by the hearing aid can be obtained by measuring the displacement at the tip of the plastic chip of the artificial middle ear, it is unclear that whether the excitation force generated by the vibrator coil is transmitted to the stapes motion effectively. Therefore, another attempt was made to evaluate the efficiency of the hearing aid using a human temporal bone, as shown in Fig. 3.2(a).

A temporal bone, with age of 64 years was extracted from a cadaver within 48 h of death using a Schuknecht bone saw at the time of autopsy. The tympanic membrane and middle ear were inspected using an operating microscope. A simple mastoidectomy and posterior hypotympanotomy were performed and some of the mastoid portion of the facial nerve and surrounding bone were removed for a clear and wide view of the stapes footplate. The tympanic membrane, ossicles, suspensory ligaments, tensor tympani and stapedius muscle were left intact. The temporal bone was embedded in dental cement and secured in a temporal bone holder, as shown in Fig. 3.3.

The microsphere with a diameter of 4.5 μm was then placed at the center of the stapes footplate. The vibrator coil was attached to the tympanic membrane with vaseline and frequency responses of the stapes-footplate motion were measured with a laser Doppler velocimeter (HLV-1000, POLYTEC) when a sound pressure of 60 dB SPL was applied to the microphone of the hearing aid by an earphone (83-13A/024, TIBBETS INDUSTRIES). Next, the hearing aid was removed and the frequency response of the stapes footplate was measured when a constant sound pressure of 80 dB SPL was applied to the tympanic membrane by the earphone directly as shown in Fig. 3.2(b).

Effect of the shape of the Ne-Fe-B magnet on the excitation force was also evaluated. After the experiments explained above, the magnet was replaced by another one, which is made of the same material with diameter and thickness of 3 mm. The frequency responses of stapes footplate motion were then measured and compared with that obtained from the first magnet.

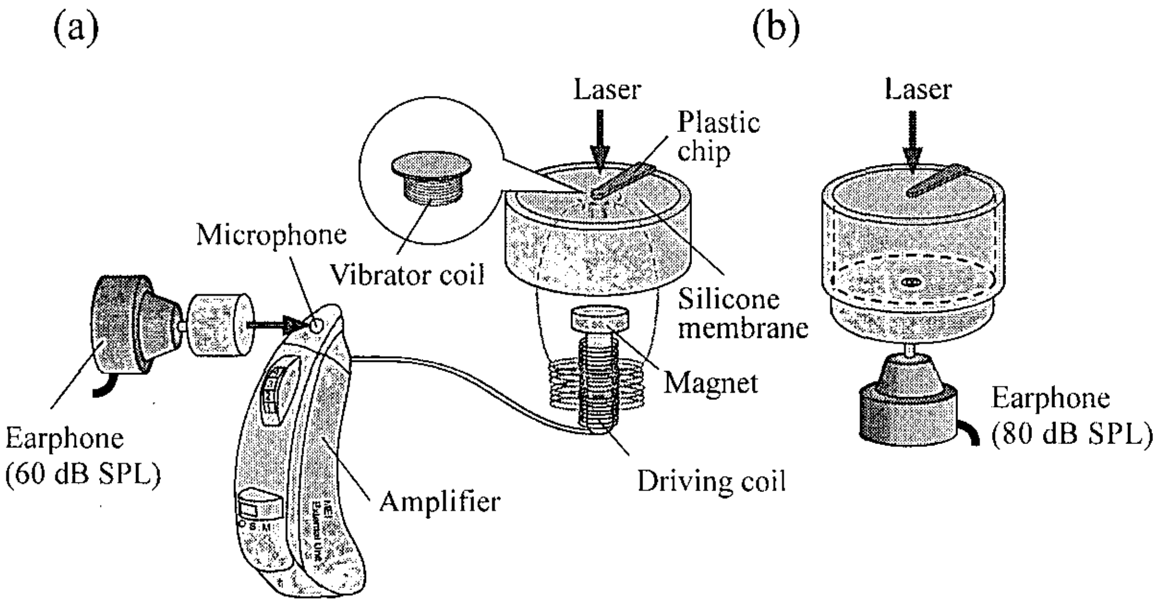

Figure 3.1. Experimental setup for measuring the displacement at the tip of the plastic chip of the artificial middle ear. (a) Experimental setup using the hearing aid. The vibrator coil was attached to the silicone membrane with oil. The distance between the magnet and the vibrator coil was 3 mm. (b) Experimental setup using the earphone. The earphone was inserted into the plastic tube and the vibration at the tip of the plastic chip was measured.

(a)

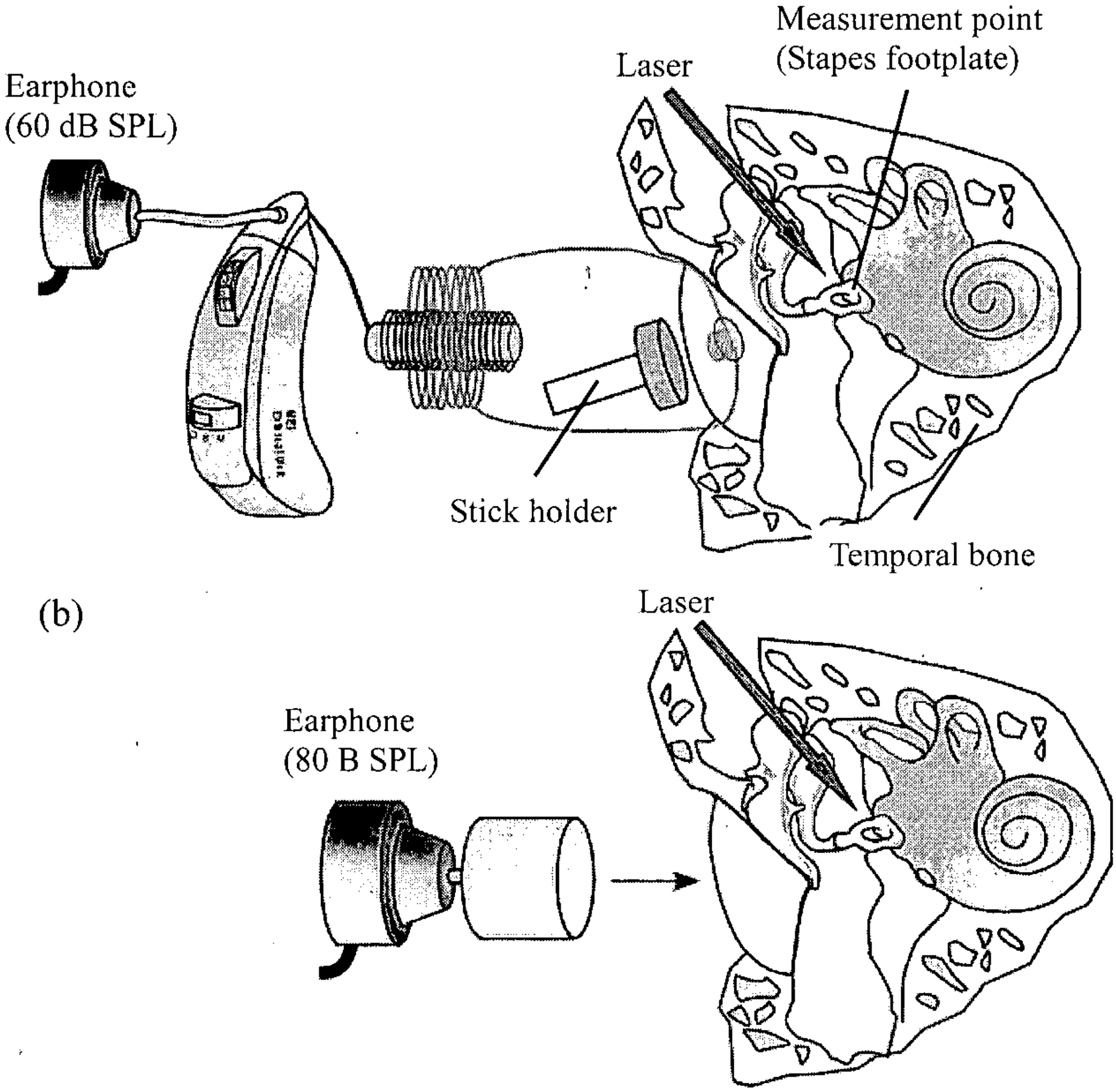

Figure 3.2. Experimental set up for the measuring the stapes-footplate motion. (a) Experimental setup using the hearing aid. After the simple mastoidectomy, the hearing aid was set and frequency responses of the displacement at the center of the stapes-footplate motion was measured with a Laser Dopper velocimeter when constant sound pressure of 60 dB was applied to the microphone of the hearing aid via tube. (b) Experimental setup using the earphone. In order to estimate the excitation force, the hearing aid was removed and the frequency responses of the stapes motion was measured when the constant sound pressure of 80 dB SPL was applied to the tympanic membrane directly by earphone.

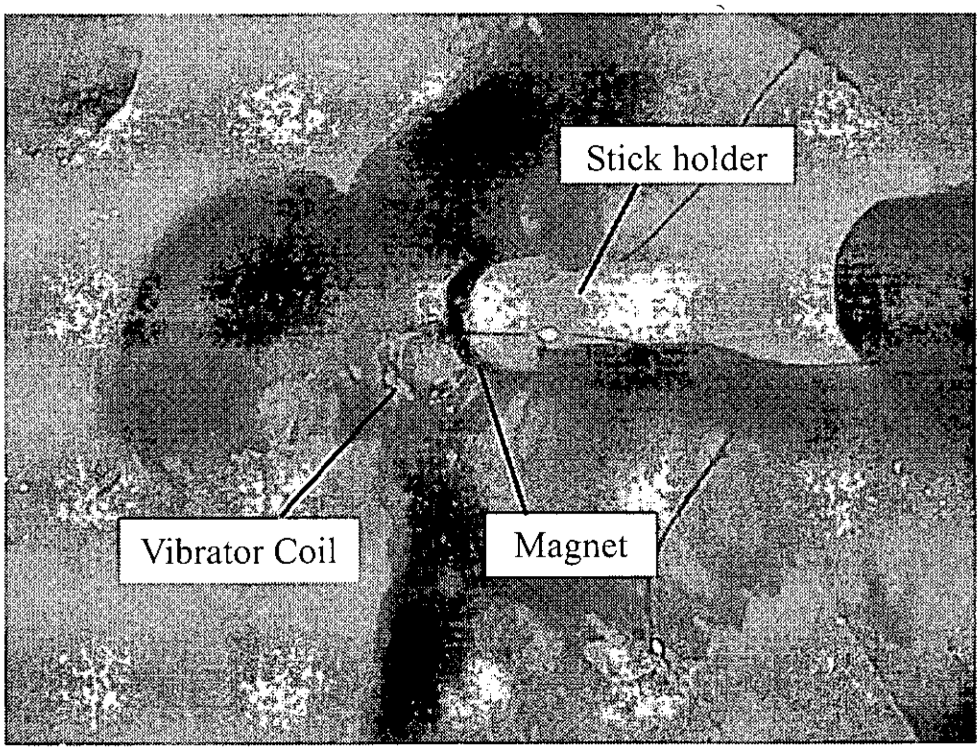

Figure 3.3. Photograph of the experimental set up for the measuring the stapes-footplate motion caused by the hearing aid. Because the external ear canal was removed in the mastoidectomy, the magnet was fixed by a stick holder and the vibrator coil was attached to the tympanic membrane with Vaseline. The distance between the magnet and the vibrator coil was 3 mm.

Chapter 4. Results

4.1. Excitation force generated by the hearing aid evaluated by using artificial middle ear

Figure 4.1(a) shows the frequency responses of the displacement at the tip of the plastic chip of the artificial middle ear caused by the hearing aid and the acoustical stimulation. The displacement excited by the hearing aid and acoustical stimulus shows large peaks at 0.5 and 0.7 kHz, respectively. The reason of this difference is considered that the vibrator coil adds mass to the artificial middle ear and the resonance frequency is shifted.

When the sound pressure of 60 dB (normal speech level) is applied to the microphone of the hearing aid, the displacement caused by the hearing aid is lager than that excited by acoustical stimulation of 80 dB SPL.

Because the acoustical stimulus level and the displacement of the membrane have a linear relationship, the equivalent sound pressure generated by the hearing aid can be calculated based on the difference between the displacement caused by the hearing aid and that by the acoustical stimulus. The equivalent sound pressure level, P_{EQ}, is represented in dB SPL as follows:

$$P_{EQ} = 80 + 20 \times \log \frac{D_{aid}}{D_{so}},$$

where D_{aid} and D_{so} are displacement at the tip of the plastic chip caused by the hearing aid and acoustical stimulus of 80 dB SPL, respectively.

The frequency response of the equivalent sound pressure is shown in Fig. 4.1(b). There is a large peak at 0.5 kHz. This is due to the difference between the resonance frequency of the artificial middle ear without a vibrator coil and that with one. When the sound pressure of 60 dB is applied to the microphone of the hearing aid, it generates an equivalent sound pressure of more than 80 dB SPL above 1 kHz. Especially above 2 kHz, the maximum excitation force generated by the hearing aid is 100 dB SPL and mean excitation force is 95 dB.

4.2. Stapes-footplate motion in human temporal bone

Figure 4.2(a) shows the frequency response of the stapes-footplate motion. The displacement of the stapes footplate is lager than that excited by acoustical stimulation of 80 dB SPL with the earphone in the frequency range of 0.5 – 5 kHz.

Based on the results in Fig. 4.2(a), an acoustical gain was calculated. In this experiment, an acoustical gain was defined as the ratio of sound pressure level generated by the hearing aid to input sound pressure level (60 dB SPL).

The frequency response of the acoustical gain obtained when the sound pressure of 60 dB is applied to the microphone of the hearing aid is shown in Fig. 4.2(b). The acoustical gain increases with an increase in frequency below 4.5 kHz except some peaks. The maximum acoustical gain was 42 dB at 4.5 kHz. However, it decreases dramatically up to 4.5 kHz.

4.3. Effect of the shape of the magnet on the excitation force

Figure 4.3(a) shows the frequency responses of the stapes footplate motion obtained when the magnet with diameter and thickness of 3 mm was used. When the sound pressure of 60 dB was applied to the microphone of the hearing aid, the displacement of the footplate excited by the hearing aid was larger than that excited by acoustic stimulation of 80 dB SPL with the earphone below 4 kHz and above 8 kHz.

Figure 4.3(b) shows the frequency responses of the acoustical gain calculated on the basis of results in Fig. 4.3(a). Almost entire frequency range, acoustical gain was more than 20 dB SPL. The maximum gain was 57 dB at 2 kHz and this value was about 15 dB larger than that obtained from smaller size of magnet. This result indicates that when the larger size of the magnet was used, the hearing aid can generate larger excitation force.

(a) (b)

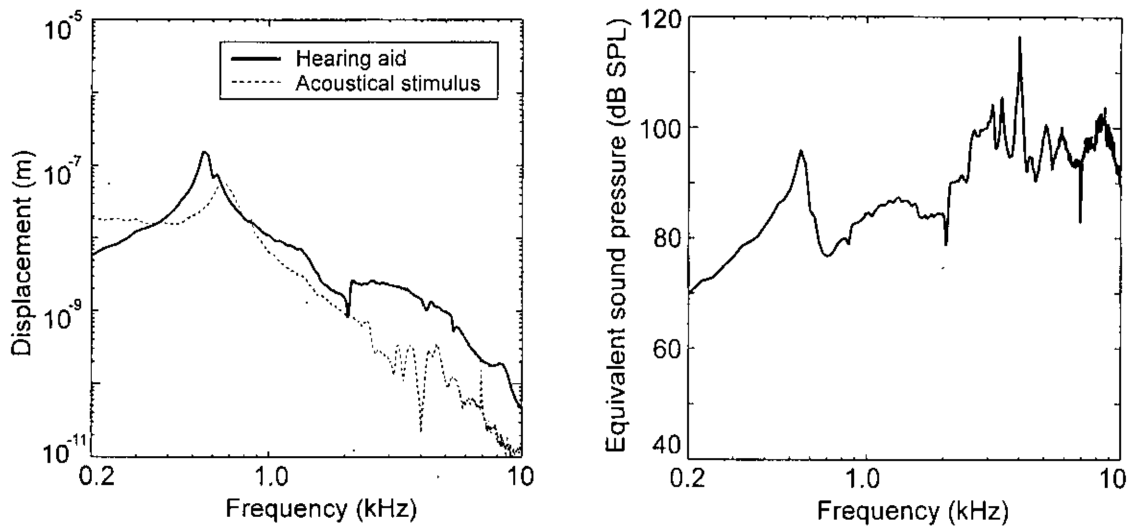

Figure 4.1. Evaluation of the excitation force generated by the hearing aid. (a) Frequency responses of the displacement at the tip of the plastic chip of the artificial middle ear. When the displacement of the artificial middle ear obtained by the hearing aid (Solid line) is larger than that obtained by the acoustical stimulation (Dotted line), this hearing aid can generate an excitation force of more than 80 dB SPL. (b) Frequency responses of the equivalent sound pressure. The maximum equivalent sound pressure generated by the hearing aid is approximately 100 dB SPL above 2 kHz.

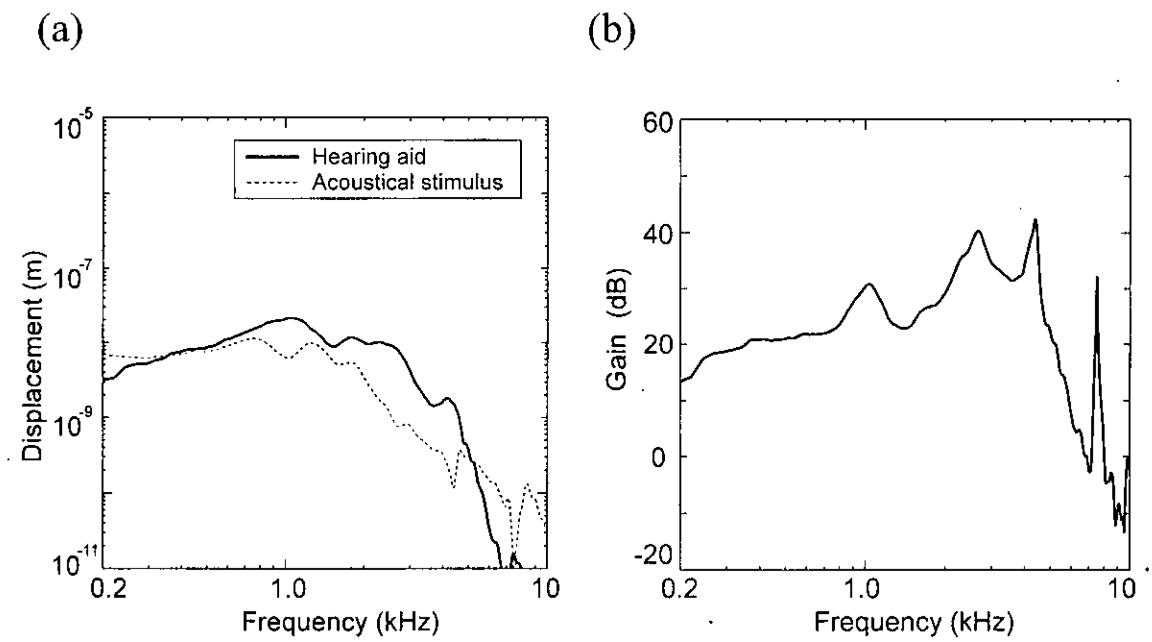

Figure 4.2. Evaluation of the efficiency of the hearing aid. (a) Frequency responses of the displacement at the stapes footplate. When the sound pressure of 60 dB is applied to the microphone of the hearing aid, the displacement of the stapes footplate is lager than that excited by acoustical stimulation of 80 dB SPL with the earphone in the frequency range of 0.5 – 5 kHz. (b) Frequency responses of the acoustical gain. The acoustical gain of the hearing aid increases with an increase in frequency below 4.5 kHz, while it decreases dramatically above 4.5 kHz.