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Original Article

Designing the Vibrating Implantable Middle Ear Transducer Using Planar Coils and Finite Element Analysis

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Abstract

Objective: The aim of this study was to develop an ideal electromagnetic vibration transducer designed for implantable middle ear device with the following characteristics: small in size and high-energy efficiency.
Materials and Methods: In order to find the output of electromagnetic force and to predict the frequency-amplitude characteristics, a finite element middle ear biomechanical model was used to derive the optimal magnetic force of the actuator in this study. First, the electromagnetic transducer was created using computer-aided design. The air gap between the magnet and planar coil, input current and vibration force were calculated using finite element analysis simulation.
Results: The simulated results showed that the electromagnetic forces under 0.2 mm air gap were about 2–6 dyne depending on the layers of the planar coils under the input current of 20–60 μA. By increasing the layers of the planar coil, the electromagnetic forces can be increased.
Conclusion: This designed actuator could be used as transducers in middle ear implants. [*Tzu Chi Med J* 2008;20(3):196–200]

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1. Introduction

Hearing impairment affects about 30% of the population aged more than 65 years in Taiwan. In fact, almost half of those over the age of 75 years suffer from hearing impairment (1). With aging, there is a loss in hearing sensitivity that is progressive and produces difficulty in understanding the spoken word, particularly when background noises are present. This loss is the most common cause of adult-onset hearing loss and is due to a gradual loss of cochlea hair cells and neurons. Hearing devices have long been the principal method of treatment for sensorineural hearing impairment to improve quality of life. There are several reasons why hearing aids are not more widely used. Conventional acoustic hearing devices have many problems regardless of the sophistication of the electronic sound processor used. The position of the hearing aid in the auricle and external ear canal not only blocks sound transmission but also interferes with the normal resonances and frequency amplification (2). Further distortion occurs when the electronic signals are converted back into acoustic signals through the output receivers. Acoustic feedback is another problem with conventional hearing devices. The output from the device is picked up by the input microphone from sound leaking into it between the device and the canal [1,2]. The occlusion effect is the phenomenon of sound not being able to dissipate normally and gives rise to the annoying perception of one's own loud voice and hollow sounds. Due to the biomechanical advances in circuitry, implantable hearing devices (IHDs) have become another option [3]. An IHD could be viewed as an extension of the conventional acoustic hearing aid technology, except that one or more of the components might be implanted into the middle ear or cochlea. The potential advantages include sound enhancement and fidelity, as well as possible total concealment. IHDs may also be applicable for conductive and mixed hearing impairment, in patients not adequately amenable for reconstructive surgery. The advantages of the IHD include eliminating the receiver and directly stimulating the ossicles or the inner ear. It also improves sound fidelity compared with conventional acoustic hearing aid technology. With the output transducer implanted, the ear canal can be left unoccluded. Therefore, normal resonance is maintained and the occlusion effect of the ear mold on one's own speech is eliminated [4].

In this research, we designed an electromagnetic vibration transducer to be applied as an IHD. The characteristics of the electromagnetic vibration transducer included the small size and high-energy efficiency. In order to derive the optimal electromagnetic force and to predict the frequency-amplitude characteristics, finite element analysis (FEA) simulation was performed. FEA is a computer simulation technique used in engineering and biomechanical analyses, which employs the Ritz method of numerical analysis and minimization of variational calculus to obtain approximate solutions to systems. First, the vibration transducer was created using computer-aided design (CAD). The air gap between the magnet and the coil, input current and vibration force were controlled using FEA simulation. Finally, the electromagnetic force that was provided by this actuator was also calculated using this FEA model.

2. Materials and methods

2.1. *Design of the actuator*

A new type of electromagnetic actuator designed to be a transducer is shown in Fig. 1. The actuator mainly consists of a planar coil and a permanent magnet. The actuator can be implanted in the middle ear during surgery. To be an appropriate actuator, some design parameters such as size, vibration force, frequencies and output signals should be considered. The control parameters of the vibration actuators include the electrical input power, number of coil layers, and the size and mass of the magnet and coil. Hence, it is necessary to determine the structure, mass and size of the magnet and the layers of coil to give the maximum vibration force. The input signal and currents were loaded into the actuator. Therefore, the output signals would create a current in the coils of the actuator. The magnetic forces generated by the current in the coils should be the same toward the South and North poles of the permanent magnet. The wound coil could be composed of one directionally wound coil or two bi-directionally wound coils. Then, the current in the magnetic fields would create pushing and pulling forces in the actuator.

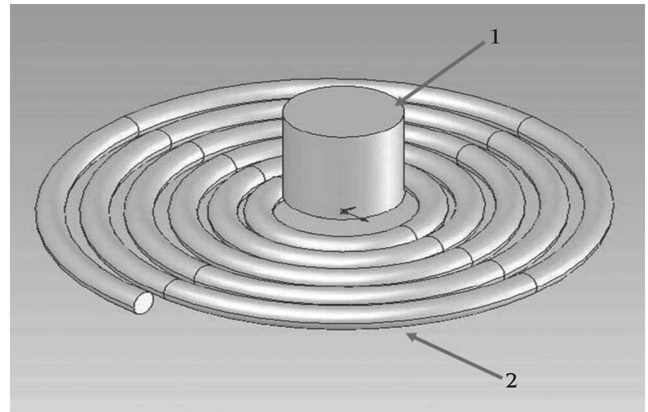

Fig. 1 — Diagram of electromagnetic vibration actuator composed of a permanent magnet (1) and planar coils (2).

2.2. *Hearing loss and force compensation*

The input sound level applied with the actuator was estimated from our previous middle ear model [3–6]. The mass of the actuator was about 80 mg, including the permanent magnet and wound coil. Electromagnetic driving parts were composed of the wound coil and a permanent magnet, as shown in Fig. 2. The diameter and the thickness of the permanent magnet were 3.0 mm and 1.5 mm, respectively, and the material was composed of neodymium-iron-boron alloy (NdFeB). The maximum magnetic flux density on the surface of the magnet was 3000 G. The distance from the magnet to the planar coil was 0.2 mm. The diameter of the planar coil was 30 μm. The outer diameter of the spiral coil was about 3.4 mm.

FEA was performed to optimize the electromagnetic force in three steps. First, the magnetic flux density around the permanent magnet was found and the electromagnetic force generated by the coil current was calculated. Second, the actuator was created using CAD and loaded to our previous middle ear model

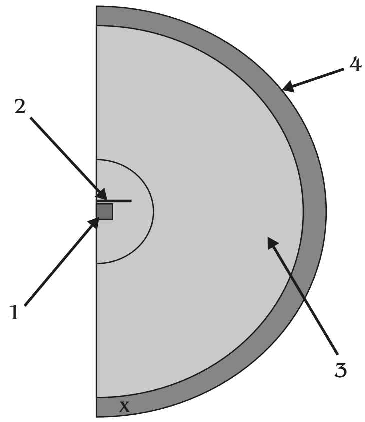

Fig. 2 — Diagram of an electromagnetic vibration actuator composed of a magnet (1) and planar coils (2). The air (3) and infinite boundaries (4) were also created using computed-aided design.

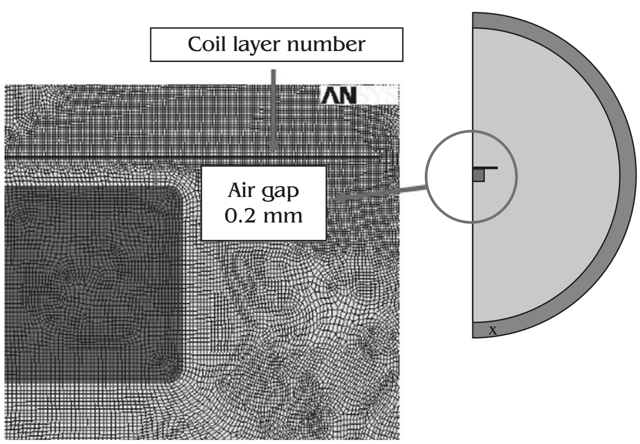

Fig. 3 — Finite element meshes and the electromagnetic driving parts: schematic diagram of the designed vibration actuator including the magnet, different layers of coils and 0.2 mm air gap.

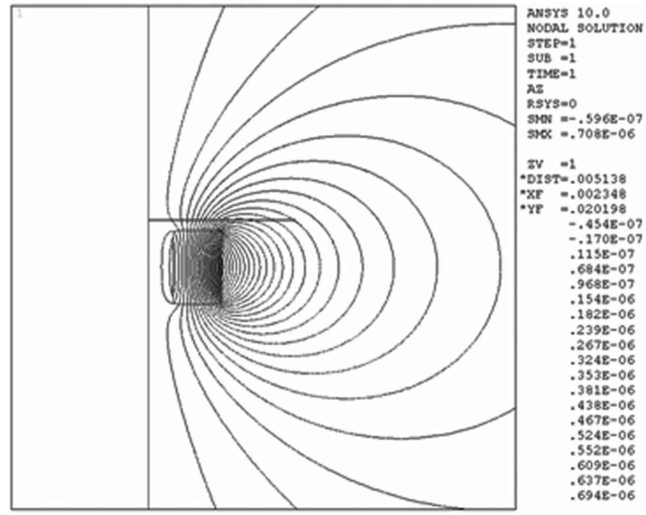

Fig. 4 — Electromagnetic driving parts: simulation results of magnetic flux distribution of the permanent magnet.

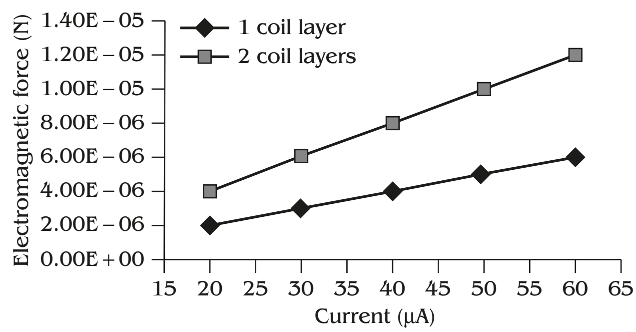

Fig. 5 — Simulation results of vibration force versus input current at the variance of turns of the coil and at the condition of 0.2 mm air gap with various input currents and different number of coil layers.

(Fig. 3). In the last step, the magnetic forces of the current in the magnetic field were calculated with numerical integration using Equation 1 (7). The magnetic flux density distribution plays a major role in the Lorentz force analysis between the coil and the permanent magnet (Fig. 4):

$$\{Fm\}=\int\{N\}(\{Bm\}\times\{Jc\})\,d(vol) \tag{1}$$

Where {Fm} is the vector matrix of the Lorentz force, {N} is the shape vector function, {Bm} is the vector matrix of magnetic flux density, and {Jc} is the vector matrix of current density. The output electromagnetic forces were calculated with 1.0 mm air gap different turns of coils and input current.

If the input sound were to be amplified for hearing to amount to 100 dB SPL, a 9.6 dyne of vibration force should be applied to the stapes using Equation 2:

$$F_s=10^{P_w/20}A_mS_sP_{ref} \tag{2}$$

Therefore, the gain was aimed to be 100 db SPL.

3. Results

The Lorentz force between the coil and magnet versus turns of the wound coil are shown in Fig. 5, with the air gap at 0.2 mm. If the magnetic vibration actuator can be fabricated by the simulation results of Fig. 5, the actuator can provide a force of 2–6 dyne at the conditions of one layer of coil and 20–60 μA. By increasing the layers of the planar coil, the electromagnetic force would be increased. When we used two layers of coils, the actuator provided a force of 4–12 dyne at the condition of 20–60 μA.

This newly designed actuator could be placed between the malleus and stapes after removing the incus, or it may be used as a clip like the Vibrant Soundbridge (VSB) to be fixed in the incus. Otherwise, the facet could be curved into a dome shape and directly connected to the stapes. The axial magnetic force generated by the actuator can directly push the ossicles as shown in Fig. 6.

4. Discussion

The inability to communicate as a result of hearing loss can be severely disabling. The principles of an electromagnetic IHD are simple (3,4). Sound enters the microphone and is transformed into an electrical signal and amplified. It is then transformed into an electromagnetic field produced by the driving coil. The field acts to vibrate the magnet, which is attached to the malleus, incus, stapes or membranous labyrinth. In 1935, Wilska was apparently the first to apply an electromagnetic actuator by placing small pieces of soft iron weighing about 10 mg on the tympanic membrane of a human subject (7). He was able to induce the perception of a pure tone generated from a variable-frequency oscillator. Report of an early implanted device was made in 1957 by Djourno et al (8). An electric coil, made of fine silver wire surrounding an iron core and insulated by plastic film, was placed in the middle ear. The patient was said to have been able to hear sounds described as whistling noises and was also able to identify a few simple spoken words.

Several teams have explored the concept of driving the ossicles electromagnetically through a coil placed in the middle ear or mastoid cavity (9–13). Because the energy transfer in the magnetic systems is inversely proportional to the cube of the distance between the magnet and the coil, the objective is to place the device as close and invariable an interface distance as possible. The transducer should ideally permit the utilization of the smallest yet strongest magnet available to avoid overloading the ossicular chain and to optimize the distance, stability and orientation of the wound coil. In cases of a conductive or mixed loss with a damaged middle ear transmission system and a mobile stapes, placement of a magnet on the stapes or footplate will bypass the conductive component, which is a definite advantage. Goode believed that an implantable device has advantages over conventional aids in many patients with sensorineural hearing loss because the implanted aid bypasses the acoustic coupling between the receiver, the ear drum and the ossicles (14). The majority of ears have a relatively smooth frequency response. Certainly, there is no question that the middle ear is a very important part of the hearing pathway even if there is no conductive hearing loss. At high sound intensity, the middle ear muscles have the function of smoothing out the peaks and valleys by damping ossicular vibration. It is well known that the middle ear muscles have essentially no acoustic function in the human above 1000 Hz (14). The hypothesis that hearing-impaired patients with markedly irregular middle ear frequency responses will do much better with implantable transducers than those with relatively smooth frequency responses might be reasonable (14).

In this study, an electromagnetic vibration actuator that can leave the ear canal open was created and tested using our previous middle ear finite model. Optimal size and shape were generated and proved using FEA calculation. The configuration of the actuator has the potential to offer three principal advantages over conventional hearing aids. These merits include elimination of feedback and occlusion,

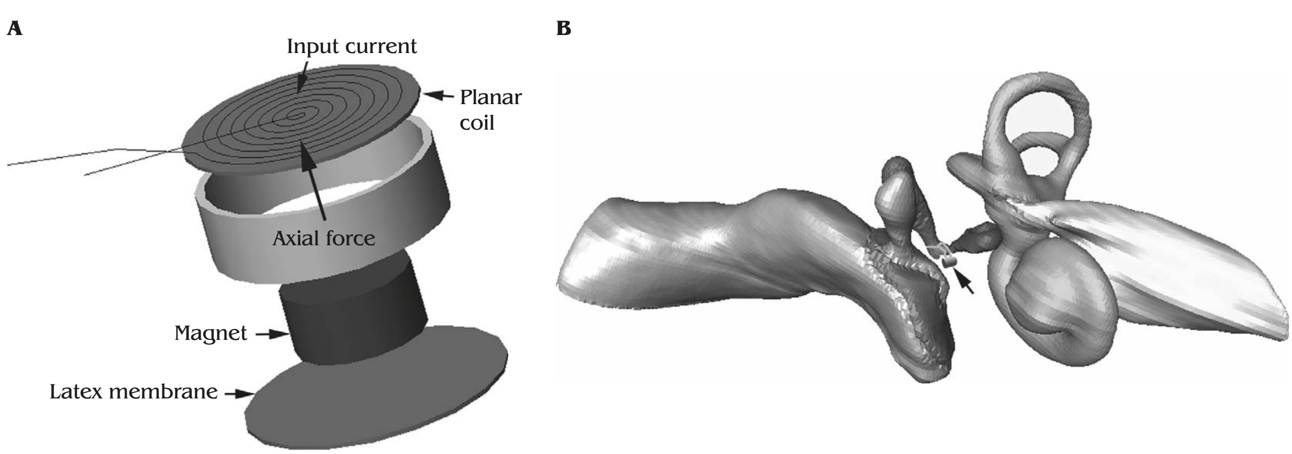

Fig. 6 — (A) Diagram of the electromagnetic vibration actuator composed of a permanent magnet and planar coils. (B) The new actuator could be attached to the incus by a clip or wire (arrow).

increased reliability and stable functional gain. In addition, the open ear canal provides a pathway for retention of normal resonances.

Regarding middle ear implants, the VSB is the only commercially available implantable hearing aid. The VSB requires a posterior tympanotomy and opening of the facial recess. Through the facial recess, the floating mass transducer is clipped to the long process of the incus. In our designed actuator, the floating mass could be implanted on the malleus, incus or stapes through posterior tympanotomy. The actual size of this newly designed actuator could be reduced using photolithography. An IHD could provide better sound and speech reception, better cosmesis or both in comparison with the conventional acoustic hearing aid. To be utilized as an IHD, the actuator should be non-irritating, biocompatible, and have a long life expectancy. The implantable components should be easily removable if the results are unsatisfactory. The preliminary results of our studies suggest that adequate amplification can be achieved from this actuator. In the future, we aim to use the load of the CAD transducer model with our developed middle ear biomechanical model, as well as fabricate the actuator and perform *in vitro* experimental testing of the actuator using the temporal bone.

Acknowledgments

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References

1. Kim HH, Barrs DM. Hearing aids: a review of what's new. *Otolaryngol Head Neck Surg* 2006;134:1043–50.
2. Perkins P. Earlens tympanic contact transducer: a new method of sound transduction to the human ear. *Otolaryngol Head Neck Surg* 1996;114:720–8.
3. Maniglia AJ. State of the art on the development of the implantable hearing device for partial hearing loss. *Otolaryngol Clin North Am* 1996;29:225–43.
4. Hüttenbrink KB. Current status and critical reflections on implantable hearing aids. *Am J Otol* 1999;20:409–15.
5. Lee CF, Chen PR, Lee WJ, Chen JH, Liu TC. Three-dimensional reconstruction and modeling of middle ear biomechanics by high-resolution computed tomography and finite element analysis. *Laryngoscope* 2006;116:711–6.
6. Lee CF, Hsu LP, Chen PR, et al. Biomechanical modeling and design optimization of cartilage myringoplasty using finite element analysis. *Audiol Neurootol* 2006;11:380–8.
7. Wilska A. Ein method zur bestimmung der horsch wellenamplituden des trammel fells bei verschieden frequenzen. *Skand Arch Physiol* 1935;72:161.
8. Djourno A, Eyries C, Vallancian B. Electric excitation of the cochlear nerve in man by induction at a distance with the aid of micro-coil included in the fixture. *CR Seances Soc Biol Fil* 1957;151:423.
9. Fredrickson JM, Coticchia JM, Khosla S. Ongoing investigations into an implantable electromagnetic hearing aid for moderate to severe sensorineural hearing loss. *Otolaryngol Clin North Am* 1995;28:107–20.
10. Hough J, Vernon J, Johnson JB, Dormer K, Himelick T. Experiences with implantable hearing devices and a presentation of a new device. *Ann Otol Rhinol Laryngol* 1986;95:60–5.
11. Maniglia A, Ko W, Rosenbaum M, et al. A contactless electromagnetic implantable middle ear device of the ossicular stimulating type. *Ear Nose Throat J* 1994;73:78–80.
12. Maniglia AJ, Ko WH, Rosenbaum M, et al. Contactless semi-implantable electromagnetic middle ear device for the treatment of sensorineural hearing loss: short-term and long-term animal experiments. *Otolaryngol Clin North Am* 1995;28:121–40.
13. Maniglia A, Ko W, Zhang R, et al. Electromagnetic implantable middle device of the ossicular-stimulating type: principle, designs and experiments. *Ann Otol Rhinol Laryngol Suppl* 1988;136:1–16.
14. Goode RL. Current status and future of implantable electromagnetic hearing aids. *Otolaryngol Clin North Am* 1995;28:141–6.