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Designing the Actuator of Hearing Aid Using Spiral Coils and Finite Element Analysis

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Abstract

Objective: An electromagnetic vibration transducer was developed to be used as a hearing aid.
Materials and Methods: This type of electromagnetic transducer should have the following characteristics: small in size, high-energy efficiency and suitable frequency bandwidth. 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 a computer-aided design. The air gap between the magnet and coil, input current and vibration force were calculated using finite element analysis simulation.
Results: Simulated results showed that electromagnetic forces under 1.0 mm air gap was about 5–25 dyne depending on the turns of coils and input current of the actuator.
Conclusion: The designed actuator could be used as transducers for middle ear implants or bone anchors in hearing aids. [*Tzu Chi Med J* 2008;20(2):125–129]

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1. Introduction

Hearing impairment affects about 30% of the population older than 65 years in Taiwan. If fact, almost half of those over the age of 75 are hard of hearing [1]. Hearing devices have long been the principal method of treatment and alleviation for sensorineural hearing impairment. Conventional acoustic hearing devices have many problems regardless of the type of electronic sound processor used. The position of hearing aids in the auricle and external ear canals not only blocks sound transmission but also interferes with 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]. Occlusion effect is the phenomenon that sound does not dissipate normally and gives rise to the annoying perception of one's own loud voice and hollow sounds. Due to biomechanic advances in circuitry, the idea of an implantable hearing device (IHD) has become another option (3). An IHD could be viewed as an extension of conventional acoustic hearing aid technology, except that one or more of the components might be implanted in the middle ear or cochlea. The potential merits include sound enhancement and fidelity, as well as possible total concealment. IHDs may also be used for conductive and mixed hearing loss that has not been adequately corrected using reconstructive surgery. The advantages of the implantable device include directly eliminating the receiver and driving the ossicles or the inner ear. An implantable device also improves sound fidelity compared with conventional acoustic hearing aid technology. The ear canal can be left open when an output transducer is implanted. Therefore, the normal resonance is maintained and the occlusion effect on one's own speech is eliminated (4).

In this study, an electromagnetic vibration transducer was developed to be an IHD. This type of electromagnetic vibration transducer must be small in size, and have high-energy efficiency and suitable frequency bandwidth. 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 analysis. The technique employs the Ritz method of numerical analysis and minimization of variational calculus to obtain approximate solution to systems. First of all, 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 by FEA simulation. Then, the new coupled tympanic membrane-transducer complex was loaded to our previous three-dimensional biomechanical model of the middle ear (5–7). In addition, the gain and frequency response curves that can be provided by this actuator were calculated using this FEA model. The model-predicted umbo and stapes displacements were close to the bounds of the experimental curves of the experimental data of Nishihara and Goode (8), Huber et al (9) and Gan et al (10) across the frequency range of 100–8000 Hz.

2. Materials and methods

2.1. Design of the actuator

A new type of opto-electromagnetic actuator designed to be a transducer is shown in Fig. 1. The actuator mainly consists of a wound coil and a permanent magnet. The actuator can be implanted in the middle ear using surgery. To be an actuator, some design parameters such as size, vibration force, frequencies and output signals should be considered. The control parameters of the vibration actuators are the electrical input power, coil turns, and size and mass of the magnet and coil. Hence, it is necessary to determine the structure, mass and size of the magnet, and turns of the coil to give the largest vibration force. The input signal and currents are loaded to the actuator. Therefore, the output signals will create currents in the coils of the actuator. The magnetic forces generated by the currents in the coils should be the same toward the S and N poles as the permanent magnet. The wound coil can be composed of one directional or two bi-directionally wound coils. Then, the currents in the magnetic fields will create push and pull forces in the actuator.

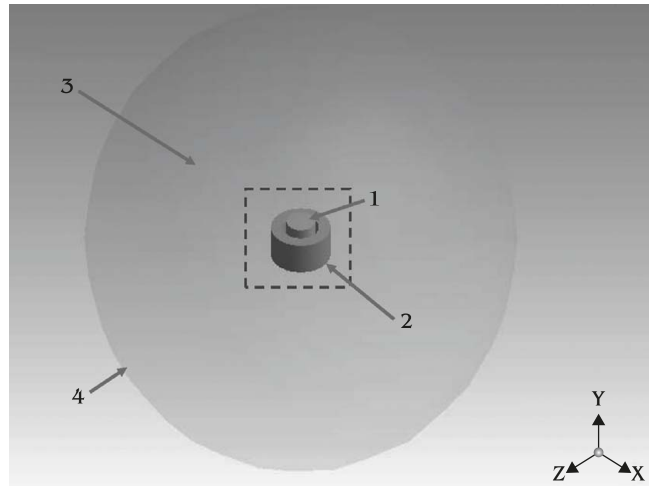

Fig. 1 — Diagram of the electromagnetic vibration actuator composed of magnet (1) and wound coils (2). The air (3) and infinite boundaries (4) were also created using computer-aided design.

2.2. Hearing loss and force compensation

Input sound levels applied with the actuator were estimated from our previous middle ear model (3–6). The mass of actuator was about 80 mg, including the permanent magnet and wound coil. The gain was aimed at 20–30 dB constantly. Electromagnetic driving parts were composed of a wound coil and a permanent magnet, as shown in Fig. 2. The diameter and thickness of the permanent magnet were 3.0 and 1.5 mm, respectively, and it was made of NdFeB. The maximum magnetic flux density on the surface of the magnet was 3000 G. The distance from the magnet to the coil was 1.0 mm. The diameter of the wound coil was 30 μm. 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

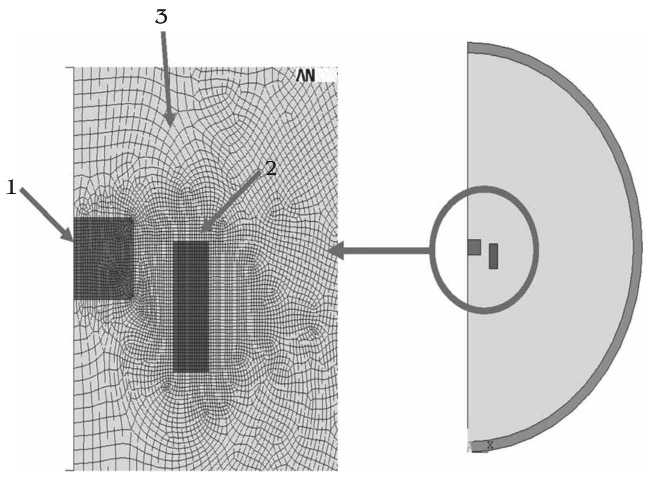

Fig. 2 — Finite element meshes and the electromagnetic driving parts: schematic diagram of the designed vibration actuator including magnet (1), wound coils (2) and air (3).

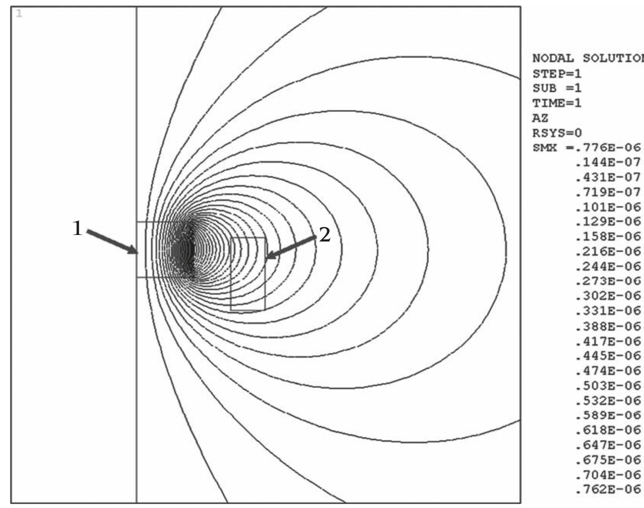

Fig. 3 — Electromagnetic driving parts. Simulation results of magnetic flux distribution of permanent magnet: schematic diagram of the designed vibration actuator including magnet (1) and wound coils (2).

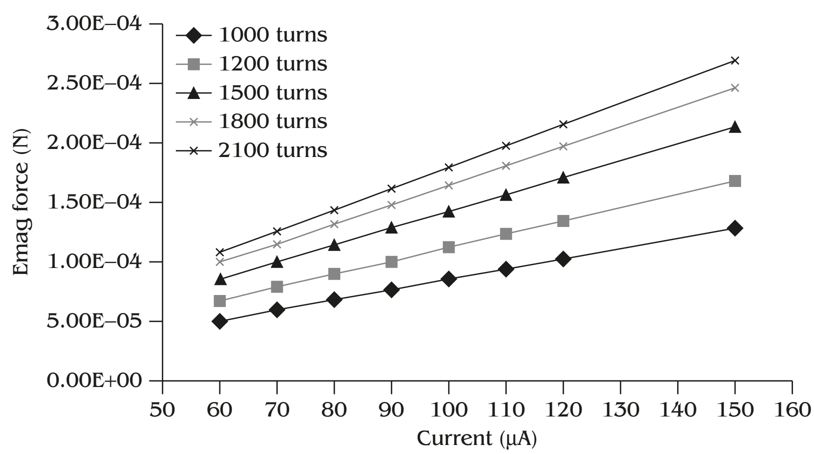

Fig. 4 — Simulation results of the vibration force versus the input currents at the variance of turns of coil and at the condition of 1.0 mm of air gap with variation input currents and different numbers of coils.

was calculated. Second, the actuator was created using CAD and loaded to our previous middle ear model. Third, the magnetic force could be calculated using the finite element method.

Equation (1) [11] shows that the magnetic forces of the current in the magnetic field can be calculated using numerical integration. Magnetic flux density distribution is a major role of the Lorentz force analysis between the coil and the permanent magnet and is demonstrated in Fig. 3:

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

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

3. Results

The Lorentz force between the coil and the magnet versus the turns of the wound coil are given in Fig. 4 in the condition of an air gap of 1.0 mm. If the magnetic vibration actuator can be fabricated using the simulation results of Fig. 4, the actuator can provide a force of 5–25 dyn under the conditions of 1000–2200 turns of the coil and 60–150 μA. To harvest the maximum magnetic force of the actuator, an aluminum ring with an inner diameter of 3.4 mm, outer diameter of 4.6 mm and height of 1.0 mm should be manufactured.

4. Discussion

The inability to communicate as a result of hearing loss can be severely disabling. The principle of the electromagnetic IHD is 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 (12) 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. He was able to induce the perception of a pure tone generated from a variable-frequency oscillator.

Several teams have explored the concept of driving the ossicles electromagnetically through a coil placed in the middle ear or mastoid cavity (13–17). Because the energy transfer in magnetic systems is inversely proportional to the cube of the distance between the magnet and the coil, the objective of the device is as close and invariable an interface distance as possible. The transducer should ideally permit the utilization of as small and strong a magnet as possible to avoid overloading the ossicular chain and to optimize the distance, stability and orientation of the wound coil.

In this study, an electromagnetic vibration actuator which leaves the ear canal open was created and tested using a middle ear finite model. Optimal size and shape were generated and proved using FE 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, increased reliability, and stable functional gain. In addition, the open ear canal provides a pathway for retention of normal resonances. Regarding the middle ear implants, the Vibrant Soundbridge (VSB) is the only commercially available implantable hearing aid. However, the VSB requires a posterior tympanotomy and opening the facial recess. Through the facial recess, the floating mass transducer is clipped to the long process of the incus. In our actuator design, the floating mass could be implanted in the malleus, incus and stapes. An IHD could provide better sound and speech reception, better cosmesis or both compared with conventional acoustic hearing aids. To be an IHD, the actuator should be nonirritating, biocompatible, and have a long life expectancy. The implantable components should be easily removable if the patient has unsatisfactory results. Preliminary studies suggest that adequate amplification can be achieved from our actuator. In the future, we aim to improve the load of the CAD transducer model to our developed middle ear biomechanical model, fabrication of the actuator and *in vitro* experimental testing of this actuator using the temporal bone.

Acknowledgments

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