8.2. Maximum output and acoustical gain of the hearing aid

When an electrical current of 80 mA flowed into the driving coil, the transducer produced an equivalent sound pressure of 93 - 106 dB SPL. Generally, hearing aids must amplify the acoustical input signal so as to satisfy the acoustical gain which is needed for the hearing aid wearer. Gain is the amount of amplification provided by the hearing aid, i.e., how much power it adds to the environmental signal. Acoustical gain of the hearing aid is determined on the basis of the most comfortable level (MCL) of the hearing aid wearer. Acoustical gain of a hearing aid can be approximately written as

$$HL / 2 \text{ [dB]}, \tag{8.1}$$

where HL is the hearing level, which means how bad the patient's own sensitivity for hearing is compared with that of normal subjects. For example, an adequate acoustical gain of the hearing aid for a patient with a hearing level of 70 dB HL is 35 dB. Such patients can understand only very loud voices at the patient's ear. Therefore, as the sound pressure level of the speech range is approximately 60 dB SPL, it is necessary for a hearing aid to output a sound pressure of more than 95 dB SPL.

Additionally, the output of a hearing aid must not exceed the uncomfortable loudness level (UCL) of the hearing aid wearer. This sound pressure level is called maximum output. Maximum output of the hearing aid is approximately given by

$$100 + HL / 4 \text{ [dB SPL]}. \tag{8.2}$$

For example, the maximum output of a hearing aid for a patient with a hearing level of 70 dB HL is 117.5 dB SPL. Therefore, the hearing aid must generate a sound pressure of 95 -117.5 dB SPL for patients with a hearing loss of 70 dB HL.

The hatched area in Fig. 8.2 shows an adequate sound pressure level which should be generated by a hearing aid for a patient with a sensorineural hearing loss of 70 dB HL. The solid line is the equivalent sound pressure level obtained when an electrical current of 80 mA flowed into the driving coil of the newly developed transducer, as shown in Fig. 7.7. The equivalent sound pressure level is approximately included within this area. These results indicate that the transducer developed in this study can possibly be used to treat patients with high-grade hearing loss up to 70 dB HL.

8.3. Comparison with other transducers

Table 8.1 shows specifications of other transducers. The piezoelectric transducer presented by Suzuki et al. (1995), mentioned in Section 3.3.1, achieved an equivalent sound pressure of 88 dB SPL at 1 kHz in the cat. However, the frequency response of the bimorph falls off at 5 kHz. Considering the audible frequency, hearing aids must achieve a high equivalent sound pressure even at 10 kHz. The equivalent sound pressure level of the transducer newly developed in this study is 98 dB SPL at 1 kHz, as shown in Fig. 5.7 and is 10 dB greater than that of the piezoelectric transducer by Suzuki et al.

Experiments with placement of implanted piezoelectric vibrators on the round window were performed by Dumon et al. (1993) using a guinea pig and human temporal bones. Their results demonstrated a flat frequency response for 0.25 - 8 kHz with an equivalent sound pressure of 85 - 110 dB SPL, which was approximately the same as that of our transducer.

Vibrant Soundbrige®, mentioned in Section 3.3.2, achieved an equivalent sound pressure level of 100 dB SPL in the frequency range of 1.5 - 6 kHz. However, except for this frequency range, the transducer could not obtain a satisfactory gain because of the mass of the transducer.

Heide et al. (1988) described their results in six patients with sensorineural hearing loss who used a non-implantable hearing aid with the coil placed in the ear canal. In this transducer, a samarium-cobalt magnet was attached to the umbo. The mass of the magnet was between 25 and 35 mg. This hearing aid achieved an average gain of 17.5 dB at 0.5, 1, and 2 kHz. However, the gain at 4 kHz was significantly lower, probably because of the effect of the mass on the sound transmission and the increased coil electrical impedance at high frequencies.

Perkins et al. (1996) described another approach to hold the magnet on the tympanic membrane. The magnet was attached to a silicone disk similar to a contact lens and held on the tympanic membrane by a small amount of mineral oil. The device is 3 mm in diameter and is called the Earlens®. It is driven by an induction loop with an amplifier, battery, and microphone, which is worn around the neck. This device has the advantage that the magnet can be easily removed if there are any problems. However, maximum gain was 25 dB at 2 kHz and was poor above 2 kHz.

Other electromagnetic transducers have similar problems at middle and high frequencies (> 4 kHz) (Zenner et al., 2000). Only the transducer by Frederickson et al. (1995) attains satisfactory gain even at frequencies above 10 kHz. The design of this transducer is intended for patients with moderate to severe sensorineural hearing loss. The transducer with a transmitting titanium shaft attached to the incus is implanted in the temporal bone. The linear motor converts the electrical signal into mechanical motion of the transmitting shaft, which vibrates the ossicular chain. The equivalent sound pressure is 140 dB SPL, and the frequency response is flat. However, because it is necessary to make a cavity in the temporal bone to hold the transducer, the technique of the implantation is extremely difficult.

The transducer in this study was able to produce high output even in a at high frequency range and would be easy to wear. As the frequency range of the satisfactory gain was extended up to 10 kHz, the transducer could be applied to even high-frequency hearing loss that typically occurs with increasing age. Additionally, as implantation is not necessary, the newly developed transducer would be applicable even in growing children.

8.4. Time history responses

Time history responses of the transducer and its magnitude spectra revealed an apparent similarity between the electrical input signal and the vibration caused by the transducer. On the contrary, the time history responses of the earphone had relatively larger distortions at all frequencies. It is considered that characteristics of the earphone and the shape of the external ear canal caused this larger distortion. Distortion occurs when the electronically processed signal is converted into an acoustic signal and emitted into the ear canal through the output speaker of the earphone. Additionally, interference of the acoustical signal was caused and the acoustic characteristics are affected by the narrow and complicated shape of the external ear canal. As a result, the acoustical signal might be distorted before it reaches the tympanic membrane.

In the newly developed transducer, the electronically processed signal is not converted into an acoustic signal and not affected by the shape of the external ear canal. In addition, the mass of the vibrator coil is light. These advantages enable a high fidelity response by the transducer, which has lower distortion than other transducers.

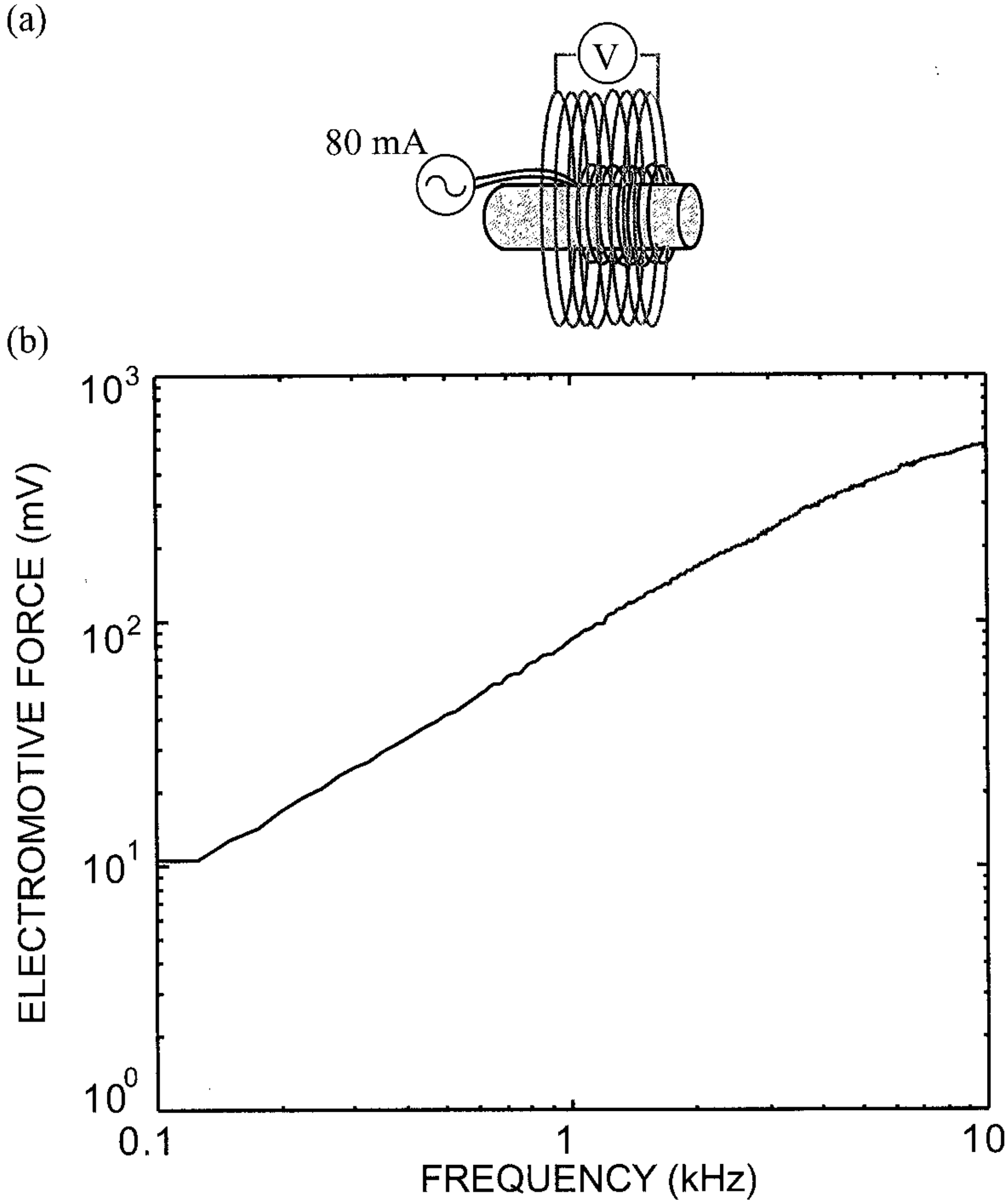

Figure 8.1. Electromotive force generated at the induction coil (open-circuit voltage). (a) Experimental setup for measurement of the electromotive force. (b) Relationship between the frequency and electromotive force. Because electromotive force generated at the induction coil increased with increasing frequency, the repulsive force which acts on the tympanic membrane as a result of magnetic interaction between the magnetic field of the magnet and vibrator coil was low.

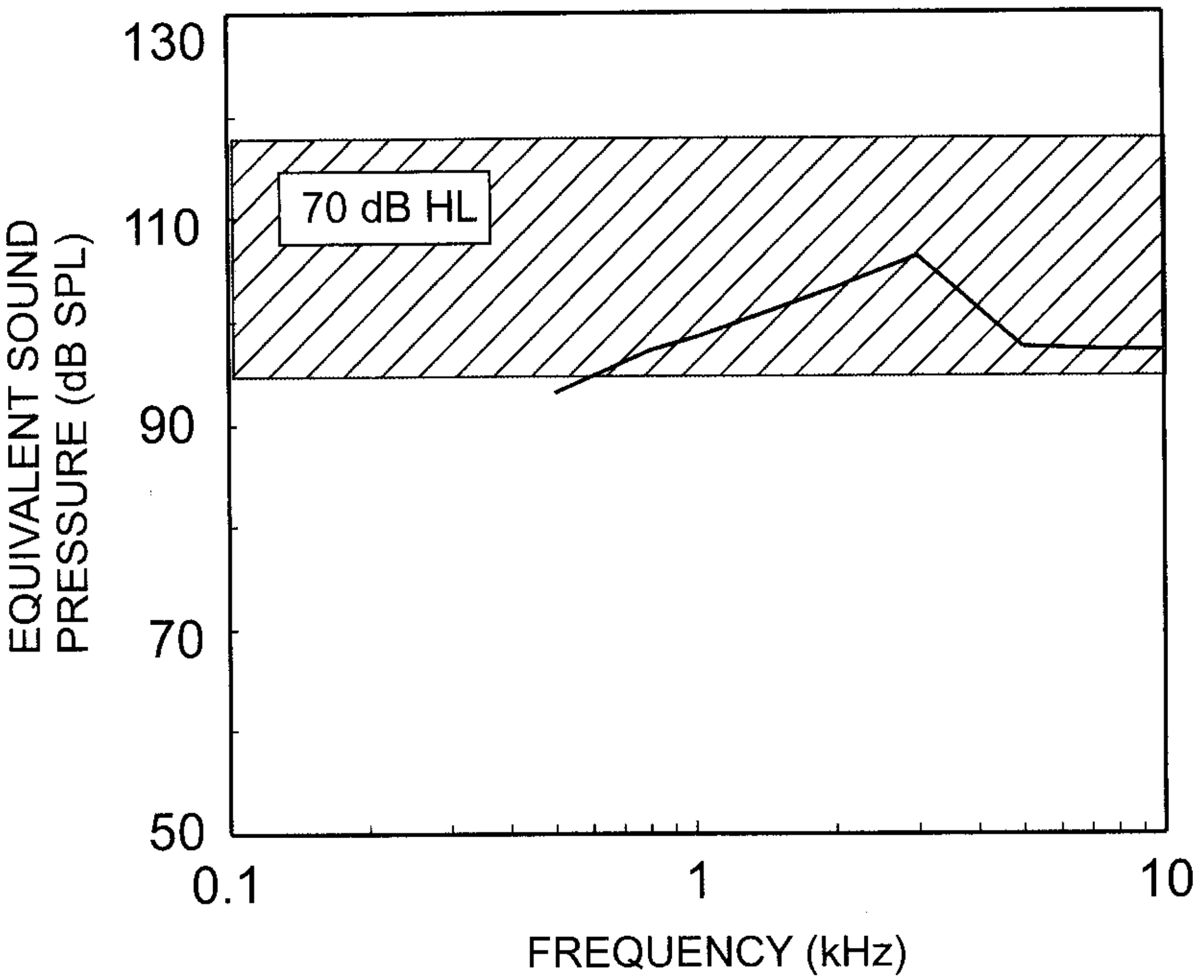

Figure 8.2. Adequate sound pressure level which should be generated by a hearing aid for patients with sensorineural hearing loss of 70 dB HL. Such a patients can understand only very loud voices at his ear. The frequency characteristics of the equivalent sound pressure level of the electromagnetic hearing transducer, as mentioned in Fig. 7.7, are included within this area.

Table 8.1. Specifications of other transducers.

Technique description	Implant or non-implant	Acoustical gain or equivalent sound pressure level	Frequency range	Subject	Reference
Our transducer	**Non-implant**	**93 -106 dB SPL**	**0.5 - 10 kHz**	**Guinea pig**	
Piezoelectric vibrator on stapes	Implant	88 dB SPL	< 5 kHz	Cat	Suzuki et al. (1995)
Piezoelectric vibrator on round window	Implant	85 - 110 dB SPL	0.25 - 8 kHz	Guinea pig	Dumon et al. (1993)
Piezoelectric vibrator on semicircular canal	Implant	———	———	———	Welling et al. (1993)
Vibrant Soundbrige® (long process of the incus)	Implant	100 dB SPL	1.5 - 6 kHz	Patients	Hough et al. (2002)
Magnet on stapes	Implant	35 dB gain	———	Cat	Maniglia et al. (1988)
Magnet on umbo	Non-implant	17.5 dB gain	< 2 kHz	Patients (n=6)	Heide et al. (1988)
Magnet on umbo (Earlens®)	Non-implant	25 dB gain	< 2 kHz	Patients	Perkins et al. (1993)
Magnet on round window	Implant	———	———	Guinea pig	Spindel et al. (1995)
Magnet on ossicular chain	Implant	60 dB gain	———	Patients (n=6)	Tos et al. (1994)
Transmitting shaft vibrated by linear motor (incus)	Implant	140 dB SPL	More than 10 kHz	Patients	Frederickson et al. (1995)

9. Conclusions

In order to develop an effective, non-invasive electromagnetic hearing aid which directly vibrates the ossicular chain via the tympanic membrane, some simple prototypes of transducers were made, and their fundamental properties were evaluated by experiments using an artificial middle ear and guinea pigs. The following conclusions can be drawn.

1. The non-implantable electromagnetic hearing transducer developed in this study is suitable for treating patients with the sensorineural hearing loss of 70 dB HL.

2. Because the newly developed transducer has flat frequency characteristics and high gain at high frequencies compared with other transducers, it can be applied to the high frequency hearing loss that typically occurs with increasing age.

3. The newly developed transducer has higher fidelity responses than those of acoustical stimulus by earphone.

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10. 研究発表

10.1. 論文発表

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