Application of MEMS to Cochlear Implants

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Abstract A cochlear implant restores some hearing by electrically stimulating residual auditory nerve fibers in the cochlea. The cochlear implant represents a major scientific and technological breakthrough and is now providing hearing for thousands of profoundly and totally deaf people around the world. In this paper, we review the present multiple-channel cochlear implant technology and explores potential applications of micro-electro-mechanical system (MEMS) technology. A new generation of electrode arrays based on the silicon micromachining technology is presented. Approaches in the use of MEMS technology for a middle ear acoustic sensor in a totally implantable prosthesis is also discussed, with key issues for its development highlighted.

INTRODUCTION

Deafness affects about 10% of the general population, and the rate increases to more than 30% for those aged over 65. Worldwide, the total number of people with some degree of hearing loss is estimated to be more than half a billion [1]. Of these, 5% are profoundly or totally deaf and gain little or no benefit from conventional hearing aids. However, a hearing prosthesis such as a cochlear implant can provide some hearing and significantly improve quality of life.

A cochlear implant consists of a surgically implanted receiver-stimulator and an externally worn speech processor. The receiver-stimulator uses an electrode array to electrically stimulate residual hearing nerve fibres in the cochlea, while the speech processor codes speech sounds into electrical stimulus and controls the receiver-stimulator via a radiofrequency link across the skin. Today, a variety of cochlear implant systems are in use around the world. Of these, the Nucleus 22-channel system is the most widely used multiple-channel cochlear implant for both adults and children. It was first developed at the University of Melbourne in the late 1970’s and subsequently commercialised by Cochlear Limited. Since 1982, it has been implanted in more than 20,000 people in over 50 countries [2].

The cochlear implant has proved to be safe and effective. However, the speech communication of cochlear implant users is still below that of normal hearing people, primarily due to the limited capacity of the electro-neural interface to convey speech information through to the auditory pathways. It is therefore necessary to improve the existing electrode array to provide more stimulation sites, better channel resolution and closer placement to the auditory nerve cells [3]. To overcome the constraints of conventional manufacturing techniques in developing these advanced electrode arrays, it may be advantageous to exploit the capability of MEMS technology, which has been extensively used in silicon microprobes for electrical stimulation and recording of nervous systems in the laboratory [4, 5].

Another possible application of MEMS technology to cochlear implants is to develop an implantable acoustic sensor for a totally implantable prosthesis. At present, the use of cochlear implant is restricted in many activities such as swimming and vigorous sport, as it is only a partially implanted system. The externally worn parts are considered by some as unsightly and are prone to damage. A totally implantable cochlear implant would be a significant improvement, with all components implanted under the skin. To achieve this, an implantable acoustic sensor needs to developed to replace the present externally worn microphone.

This paper explores some potential applications of micromachine technology to cochlear implants by reviewing examples of recent progress in this field. It will first describe the present cochlear implant technology and the communication performance of adults and children using the device. It will then explore the development of advanced electrode arrays including a micromachined flexible array and various peri-modiolar electrode arrays. Finally, possible approaches for the development of an implantable middle ear sensor using MEMS technology will be discussed.

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MULTIPLE-CHANNEL COCHLEAR IMPLANT

As shown in Fig. 1, the human ear is divided into the outer ear, middle ear and inner ear. The outer ear consists of the pinna and external auditory meatus (or ear canal), which terminates at the tympanic membrane (or eardrum). Behind the eardrum is the air-filled middle ear cavity, where three ossicles (small bones) form the ossicular chain, connecting the eardrum to the cochlea (the inner ear). In normal hearing, sound waves pass down the ear canal and vibrate the tympanic membrane and the ossicles, which transfer the airborne sound into pressure waves in the fluid-filled cochlea. The snail-shaped cochlea has three ducts, namely the scala vestibuli, scala media and scala tympani (Fig. 1). As the pressure waves propagate through the cochlea, the basilar membrane is displaced due to the pressure difference between the scala vestibuli and scala tympani. As a result, the sensory hair cells in the organ of Corti are actuated, subsequently triggering action potentials (nerve firings) in the adjacent spiral ganglion cells.

Because of the varying stiffness and shape of the basilar membrane along the length of the cochlea, the region of maximal displacement varies with frequency. Higher frequency sounds cause greatest displacement at the basal end of the cochlea, while lower pitch sounds maximally displace the basilar membrane at the apical end. Since displacement of the basilar membrane causes excitation of the auditory nerve fibres associated with that region, different groups of nerve fibres are stimulated depending on the frequency of the sound. Not only is the basilar membrane tuned in this way, but the nerve fibres themselves have a best frequency of response. According to the place theory of sound coding, then, higher frequency sounds activate nerve fibres at the basal end of the cochlea, while lower frequency sounds activate more apical fibres. This is referred to as tonotopic ordering.

Profound and total deafness is usually the result of severe damage or death of the sensory hair cells due to disease process, trauma or congenital abnormality. A cochlear implant bypasses the hair cells and directly stimulates the residual nerve fibres. The Nucleus multiple-channel cochlear implant has 22 electrodes inserted in the scala tympani. To exploit the tonotopic ordering of the cochlea, the implant assigns a fixed frequency band to each of the 22 electrode positions, and uses the site of stimulation to code the frequency. Intensity is coded by the amplitude of the stimulus, while the amplitude envelop is used to provide temporal information.

Fig. 2 illustrates the latest Nucleus-24 system developed by Cochlear Limited in collaboration with the Cooperative Research Centre (CRC) for Cochlear Implant, Speech and Hearing Research [6]. It consists of an advanced CI-24M receiver-stimulator (R/S), a directional microphone, and a body worn speech processor (Sprint™). The system also offers an alternative behind-the-ear speech processor (ESprir™), where all the external components are housed in a package worn at ear level. The speech processor has an audio pre-processor, a digital signal processor (DSP), a data encoder and a radiofrequency transmitter. In operation, the microphone output is first sent to the preamplifier and signal conditioning circuitry of the audio pre-processor. Next, the DSP analyses the received audio signals and implements the speech processing strategy by assigning the appropriate electrode, current

Figure 1  Diagrams of the human ear and cross sectional view of the cochlea: A) pinna, B) ear canal, C) eardrum, D) eustachian tube, E) ossicles, F) cochlea, G) auditory nerve, H) scala vestibuli, J) scala tympani, K) scala media, L) organ of Corti, M) basilar membrane and N) spiral ganglion.

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amplitude, and stimulus rate. These data are then coded and transmitted to the R/S via a radiofrequency (RF) link between two inductive coils. The R/S consists of a flexible electrode array, a hermetically sealed titanium capsule and a platinum coil for receiving RF signals. The electrode array has 22 platinum ring electrodes at its distal end. Each electrode is connected to a custom IC chip in the titanium capsule by one of the 22 separately insulated platinum-iridium lead wires wrapped in a silicone rubber carrier. The R/S circuitry is controlled and powered by the external speech processor to select the electrodes and set the level of stimulation.

As shown in Fig. 3, enhancements in speech processing strategies have resulted in improved speech understanding performance for users of cochlear implants [7]. The strategy (FO/F2) implemented in the first Nucleus speech processor, the WSP II, coded the fundamental frequency of voiced speech as rate of stimulation. The second formant frequency was extracted to determine the electrode of stimulation. In the second commercial speech processor, the WSP III, the FO/F1/F2 strategy was used, where both the first and second formants were coded as place of stimulation, with two electrodes stimulated non-simultaneously. The next generation speech processing strategy, the Multipeak, also used three band-pass filters to determine stimulations on the three most basal electrodes, thus providing high frequency information in addition to the FO/F1/F2. Because high frequency spectral information provides important cues for consonant recognition, the Multipeak strategy implemented in the MSP speech processor led to significant improvements in speech understanding. The most recent speech processing strategy is based on the Spectral Maxima Sound Processor (SMSP). Instead of
extracting speech features as the previous strategies, it analyses speech sounds using 16 band-pass filters and selects the 6 outputs of highest amplitude for coding and transmission. The SMSP strategy was later developed and implemented in the Spectra 22 speech processor as the SPEAK strategy, which selects up to 8 spectral maxima from 20 band-pass filters. The SPEAK strategy is used in the present Nucleus 24 system.

The improvement in user speech understanding performance is evident in Fig. 3. After almost 20 years of research and development, the average speech understanding of cochlear implant users now exceeds 80% (key words presented with context in sentences without the aid of lipreading). More than 50% of implant users can communicate on the telephone. In Australia, over 50% of children who use the implant attend mainstream schools.

DEVELOPMENT OF ADVANCED ELECTRODE ARRAYS

Micromachined Electrode Array

A micromachined electrode array intended for use in cochlear implants has been developed at the University of Michigan [8]. As shown in Fig. 4, the design of such an array is basically modelled on the multiple-channel Nucleus electrode array. The Nucleus array has 22 platinum ring electrodes at a pitch of 0.75 mm in a highly flexible silicone rubber carrier, tapered in diameter from 0.4 mm at its tip to 0.64 mm for the basal 12 electrodes. The Michigan array, on the other hand, also has 22 activated iridium oxide (IrO) stimulation sites on a flexible silicon microprobe. The width of the probe is tapered from 0.32 mm at the tip to 0.64 mm over a length of 25 mm. The circular IrO sites are 0.25 mm in diameter and also has a center-to-center spacing of 0.75 mm.

The flexible microprobe is fabricated in a 4 µm thick p++ silicon layer, which also acts as a high boron-concentration etch-stop in ethylenediamine pyrocatechol (EDP). The silicon substrate beneath the 22 gold bond pads, however, is about 12 µm thick and less flexible. The stimulation sites are formed by sputtering a 50 nm thick titanium layer and a 300 nm thick iridium layer, patterned by a lift-off step. All electrode sites are connected to the bond pads via polysilicon leads, which are heavily doped with phosphorus to lower its resistivity. A stress-compensated stack of silicon dioxide and silicon nitride are used as insulation layers. Both the dielectric layers and the polysilicon lines are deposited using low-temperature chemical vapor deposition (LPCVD) and patterned by reactive ion etching (RIE). Alternatively, refractory metal tantalum (Ta) is sputtered to form the connecting leads, which results in an order of magnitude lower in resistance than that of its polysilicon counterpart. In addition, the insulation of the Ta lines is achieved by low temperature oxide (LTO) and plasma-enhanced chemical vapor deposition (PECVD) nitride in stead of the LPCVD dielectric layers. In the final step of the processing flow, the probe structure is etched and released from the wafer using EDP.

![Figure 4 Schematic drawings of the micromachined electrode array (top) and the Nucleus electrode array (bottom)](image-url)
In comparison with the Nucleus array, the micromachined electrode array has a much smaller cross-sectional area, batch fabrication capability and better reproducibility. It is also feasible to fabricate a greater number of stimulation sites than the current 22 electrodes, to improve the spatial resolution of stimulation and the amplitude envelope for better temporal coding. A drawback of the silicon array is that the surface area of the stimulation sites is significantly smaller than that of the Nucleus array. As a result, the stimulus level is reduced due to the current density limitation. Moreover, the robustness and longevity of the microprobe are yet to be tested under normal surgical procedures and operating conditions.

Peri-Modiolar Electrode Array

At present the Nucleus electrode array tends to curl along the outer wall of the scala tympani after insertion, away from the spiral ganglion cells situated in the modiolus. To improve the resolution of frequency coding and reduce the stimulus threshold for cochlear implants, it is necessary to place the electrode array close to the modiolus to achieve more localised stimulation as shown in Fig. 5. Such a modiolar-hugging mechanism is particularly important for a micromachined electrode array because of the limited stimulus levels.

Three peri-modiolar array designs have been developed to prototype stage by the Co-operative Research Centre for Cochlear Implant, Speech and Hearing Research (Fig. 6). One approach is to fabricate the array with a curvature which fits the inner wall of the scala tympani [9]. During insertion, the pre-curved array is held straight by a custom designed introducer so that the first 8 mm of the array is inserted along a relatively straight path in the basal turn of the scala tympani. The second approach is to incorporate a layer of hydrogel material on the outer curve of the electrode array. The hydrogel material swells when it absorbs water after insertion, thereby causing the array to curl and hug the modiolus. There are many types of hydrogel materials, such as Dextran, collagen, polydiallylamine,
polyvinylalcohol and polyacrylic acid (PAA). PAA has been found to have the most suitable properties [10]. A third group of prototype arrays uses a very thin strip attached to the tip of the existing electrode array. At the end of the insertion procedure, slight force is applied to the strip, which in turn bends the array towards the modiolus. Further in vivo studies are being carried out to evaluate the robustness and efficacy of each design concept. A novel feature proposed for the micromachined array is to incorporate an array of polysilicon strain gauges on the silicon probe to obtain in situ measurements of its curvature in the scala tympani. This feature would allow regular monitoring of the electrode position and may eventually lead to an actively controlled peri-modiolar array [8].

**IMPLANTABLE ACOUSTIC SENSORS**

An implantable acoustic sensor is a critical component in a totally implantable cochlear implant system. There have been a number of reports on the use of subcutaneous microphones by researchers in the field of implantable hearing aids [11-13]. The idea is to encapsulate a conventional microphone in a biocompatible package and implant it under the skin of the ear canal. Such a subcutaneous microphone, however, is likely to cause necrosis of the overlying skin and consequently protrusion of the device. An alternative method is to implant the microphone under the thinner skin behind the ear. This, however, would incur a loss of microphone sensitivity. The acoustic shadow effect of the head and pinna would also affect the microphone directionality, resulting in poorer performance in noise. In addition, subcutaneous microphones may also pick up body noises due to the bone conduction of the skull.

To overcome these problems, one approach is to implant a sensor in the middle ear to pick up the vibrations of the ossicular chain directly. Such a middle ear sensor integrates with the outer and middle ear structure and therefore exploits the acoustic amplification and filtering functions of the ear. Furthermore, as the sensor is placed in the middle ear cavity it is well protected from external impact.

One approach proposed for a middle ear sensor was to use a piezoelectric bimorph cantilever as an electromechanical transducer [14]. The bimorph cantilever is made of lead zirconium titanate piezoceramic (PZT). It weighs 20 mg and measures 7 mm in length, 1 mm in width, and 0.4 mm in thickness. In a similar structural design as a piezoelectric microphone, one end of the PZT cantilever is rigidly fixed on the skull with the free end connected onto the malleus (the ossicle attached to the eardrum as shown in Fig. 7). To ensure its biocompatibility, the PZT cantilever was sealed with a 0.1 mm thick layer of fluorocarbon resin, while the tip of the cantilever touching the malleus was made of apatite ceramic. The peak-to-peak displacement sensitivity in human temporal bones was measured to be -59 dB re 1 V/Pa at 1 kHz with a resonant frequency at about 2 kHz. Although these results were promising, no progress has been made with a prototype prosthesis for human trials.

A major drawback of the PZT cantilever sensor is that it requires intricate surgical procedures to assemble the device in the middle ear, where two fixation steps are involved to position and align both ends of the cantilever with a high degree of precision. To simplify the surgical technique, a more elegant approach is to attach an accelerometer onto one of the ossicles so that only a single coupling step is required [15] (Fig. 7). Such a design concept is based on the inertial transfer of vibration energy instead of relying on the relative displacement force between the ossicular chain and the wall of the middle ear.

Since the size and weight are critical, miniaturised silicon accelerometers are well suited for such a middle ear sensor. A batch-fabricated silicon accelerometer, based on the piezoresistive sensing technology, has been described in the literature [16]. The device package measured 3 mm in length,
2 mm in width, and 0.6 mm in thickness, weighting only 20 mg. As shown in Fig. 7, the accelerometer was fabricated in a three layer glass-silicon-glass structure. The sensing element of acceleration forces was formed in the centre layer of silicon, where a seismic mass was supported by a thin silicon cantilever beam in the middle of a 200 μm thick rim. An anodic bonding was performed to assemble the top and bottom glass layers and to achieve a hermetic seal between the silicon and glass surfaces. The accelerometer could detect accelerations from 0.001 to 50 g over a 100 Hz bandwidth. By modifying the design parameters of the silicon structure, the sensing range could be varied readily over several orders of magnitude.

Such a silicon accelerometer has many desirable features in terms of the size, weight, and biocompatibility, and would offer many advantages as an acoustic sensor for a totally implantable cochlear implant. The middle ear cavity has an overall capacity of about 2 cm$^3$ and measures 15 mm in height and about 4 mm in depth. An implant device with dimensions less than 3 mm should present little difficulty for surgical implantation, particularly where a single coupling procedure is performed. In addition, it has been found that the attachment of a mass of up to 20 mg on the ossicular chain causes a minimal loss of less than 3 dB in eardrum displacement [17]. In contrast, the vibration amplitude of the eardrum with an input of 80 dB SPL is 0.045 μm at a mean resonant frequency of 1 kHz. The frequency response is relatively flat from 140 Hz to 1 kHz. Above 1 kHz, however, there is a roll-off of -8.25 dB/octave [18]. As a result, the major technical issues are concerned with the sensitivity, dynamic range and frequency response, which are largely dependent on the dynamic response of the ossicular chain. Considerable effort is therefore needed to develop a workable middle ear sensor using this approach.

CONCLUSION

In this paper, we have reviewed cochlear implant technology and potential applications of MEMS technology in this field. While considerable progress has been made with the multiple-channel Nucleus system in the last twenty years, further improvement in speech understanding performance is still required. MEMS technology has the potential to be applied in the development of new generation electrode arrays which have a greater number of stimulation sites lying closer to the neural elements in the modiolus. This should lead to a significant improvement in information transfer across the electro-neural interface in the cochlea. Further, we have also discussed the use of micromachined input transducers in the development of a totally implantable cochlear implant. Although considerable effort is still required, it is clear that MEMS is providing an alternative to the conventional manufacturing techniques for future advances.

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