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Micromechanical Resonator Array for an Implantable Bionic Ear

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High-density microelectrode arrays · Speech processor-based cochlear implants · Electrical transduction

Abstract
In this paper we report on a multiresonant transducer that may be used to replace a traditional speech processor in cochlear implant applications. The transducer, made from an array of micromachined polymer resonators, is capable of passively splitting sound into its frequency sub-bands without the need for analog-to-digital conversion and subsequent digital processing. Since all bands are mechanically filtered in parallel, there is low latency in the output signals. The simplicity of the device, high channel capability, low power requirements, and small form factor (less than 1 cm) make it a good candidate for a completely implantable bionic ear device.

Introduction

It has long been the dream of neuroscience to develop implantable prosthetics that can repair or replace the functions of high order sensory organs such as hearing or vision. Since 1800, when Volta first placed 50 V electrodes in his ears and described hearing 'a sound, or rather noise in the ear' [Volta, 1800], medical technology has endeavored to restore hearing by direct electrical stimulation of the auditory nerve. Persons with bilateral acoustic tumors or an absence or destruction of the auditory nerve may receive auditory brainstem implants, which directly stimulate one of the auditory processing centers of the brainstem, bypassing the cochlea and auditory nerve. For profoundly deaf patients with intact auditory nerves, the best technology available for restoring hearing is the cochlear implant (CI). CIs use small implanted electrode arrays that directly stimulate the auditory nerve through the cochlea [Shannon and Otto, 1990; Rauschecker and Shannon, 2002]. These devices were proposed as early as 1969 [Simmons, 1969]. Starting with a single-electrode commercial device in 1984, CIs have advanced to multielectrode devices that can enable thousands of patients to hear and recognize speech. The number of CI users exceeds 60000 worldwide, and the number is growing exponentially [Zeng, 2004].

Direct stimulation of the auditory nerve can produce functional hearing, and hearing quality correlates with the number of stimulating electrodes. Early single-electrode devices provided essentially no open-set speech recognition except in a few subjects. Most modern CIs use multielectrode implants, which have been developed to

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take advantage of the so-called tonotopic organization in the cochlea, namely, the apical part of the cochlea encodes low frequencies while the basal part encodes high frequencies. These implants, therefore, all have implemented a bank of filters to divide speech into different frequency bands, but they differ significantly in their processing strategies to extract, encode, and deliver the right features. Current CI technology can provide 22 electrodes per implant, as in the Nucleus 24 model, available from Cochlear (Cochlear, Lane Cove, Australia).

Although having flexible programmability [McDermott, 1998; Zeng, 2004], hearing enhancement devices based on digital signal processor technology are expensive, costing typically \$30000 for CIs, and require relatively large and expensive microelectronic chipsets that consume large amounts of power, typically 50–750 mW for a CI. Consequently, the devices require large body-worn battery packs and accessories to produce the electrical signals needed for the deaf to hear. Furthermore, the battery life is limited to less than a week or just a few hours in many cases, requiring frequent recharging of the devices. The use of digital signal processors introduces latency in the audio signal of up to tens of milliseconds. Since the signal must be encoded and then transmitted via a wireless connection to electronics beneath the skull, only a limited number of channels (e.g., up to 22) can be processed. The expense, inconvenience, and frequent recharging requirement of current technology means that the majority of the hearing-impaired population cannot or choose not to fully benefit from the technology [Tyler et al., 2004]. The current market penetration rate for CIs is less than about 5% [Zeng, 2004].

Apart from practical and cosmetic concerns about speech processor-based CIs, there is a concern regarding hearing quality. Most speech processor algorithms encode temporal cues about the waveform envelope to aid the patient in interpreting speech [Saunders and Kates, 1997; Loizou, 1997, 1998]. While this is effective in distinguishing the spoken word (at least for Indo-European languages in quiet conditions), it provides little help in enabling the patient to hear true musical pitch for the appreciation of music or the understanding of tonal languages [Zeng, 2004]. Recent work has indicated that such temporal-based algorithms are unlikely to succeed – the source of tone transduction is truly tonotopic in the cochlea [Oxenham et al., 2004]. Properly positioning more electrodes in the cochlea, and properly stimulating them, is the most likely means for restoring tonal sense.

An alternate approach to cochlear implants and speech coding is the ambitious goal of building an artificial cochlea that truly mimics the behavior of the natural cochlea. Such a device could be used for research aid for understanding cochleas (e.g., for developing mathematical models), or eventually as a front-end transducer for an electrode system.

Some progress has been made in this area already. Researchers have demonstrated the ability to build fluid-filled, 1:1 scale models of the human cochlea and have demonstrated that they respond to sound in a manner similar to the known response of the mammalian cochlea [Ohyama and Koike, 1999; Lim et al., 1999; White and Grosh, 2005]. Most recently, the team of White and Grosh [2005] demonstrated the ability to use silicon micromachining technology in order to build their cochlea model. This is significant because the batch micromachining process used to fabricate the system will allow future integration of sensing elements into the structure to economically produce low-power, micromechanical, cochlear-like sensor filters. The devices built thus far are primarily for research purposes and to aid in understanding the mechanism of the cochlea.

In this same spirit, a second type of artificial cochlea may be constructed by building a mechanical bank of resonators designed to respond in a manner similar to the cochlea. A mechanical filter bank acts in a passive way to perform sub-band filtering, reducing power requirements. Furthermore, an array of such resonators may work in parallel to produce a large number of frequency bands simultaneously, reducing latency. By controlling the shape and composition of the resonators, one may design simple to complex resonances into the system, depending on the requirements of the cochlear design. The traveling wave phenomena of the cochlea may be included by lightly coupling the resonators together. A mechanical bridge version of this was demonstrated by Haronian and MacDonald [1995]. Their design employed a large array of thin bridges micromachined in silicon with lengths that were increased exponentially. This formed an array of resonators, each with a characteristic frequency. In some cases, the spacing between bridges was small enough to couple neighboring bridges (by the viscosity of air) so that the device behaved similarly to a cochlea. In addition, the viscosity of air served to dampen the resonances so that the device exhibited low Q, a desirable feature for an artificial cochlea. Apart from a single conference paper in 1995, no other work appears to be published on this research.

Japanese researchers Tanaka et al. [1998] also demonstrated a variation of this concept by fabricating an ingenious device that they called a 'fishbone' resonator. Their device, fabricated from silicon, consisted of an array of mechanical beams connected to a single torsional beam at their centers, making it appear as a 'fishbone'. The resonators in this device were coupled by the central beam making it behave as an acoustic transmission line. This construction enabled the device to mimic a cochlea. The device was not directly instrumented – the researchers used external optical instrumentation to monitor the movement of the oscillators.

A third type of artificial cochlea can be built based on electronic circuitry designed to convert an input signal into multiple outputs that mimic the cochlea. Banks of band-pass filters have been built [Loulou, 2004], as well as the so-called 'silicon cochlea', an electronic transmission line (filter cascade) designed to mimic the cochlear function [Kuszta, 1998]. The filter cascade model seems to hold great promise. By tapping in to the cascading series of filters one can achieve a large number of outputs that appear to closely mimic the gain, filtering and dynamic range characteristics of the cochlea [Lyon and Mead, 1988, 1998; Lyon, 1998]. Moreover, such a device has been built with 117 outputs over the range of 100 Hz to 10 kHz, 61 dB dynamic range, with small size (less than 3 × 3 mm) and low power consumption (0.5 mW) [Sarpeshkar et al., 1998; Sarpeshkar, 1999]. This is a tremendous feat that may well signal the next generation of artificial cochleas.

Whether fluidic, mechanical or electrical, the development of a small, low-power, analog, multiresonator system that can mimic the cochlea would be a major step toward developing a completely implantable bionic ear that can provide true, quality hearing.

Micromachined Multiband Transducer

We are developing a low-power micromachined multiband transducer, small enough to be implanted in the head, which we believe could ultimately alleviate the need for a speech processor. Power requirements for a system using this technology could be much less than conventional systems, enabling it to be run by rechargeable, implanted battery system. By doing so, we may envision a fully implantable bionic ear that can restore human hearing. The microphone consists of an array of micromechanical resonators, each tuned to a different center frequency, and each instrumented to an individual amplifier. The output from the device is a number of independent channels, each carrying an electrical signal representing a particular frequency sub-band of the original acoustic signal. We have built and tested two versions of this device. One used optical readout [Xu et al., 2004], the second used capacitive readout.

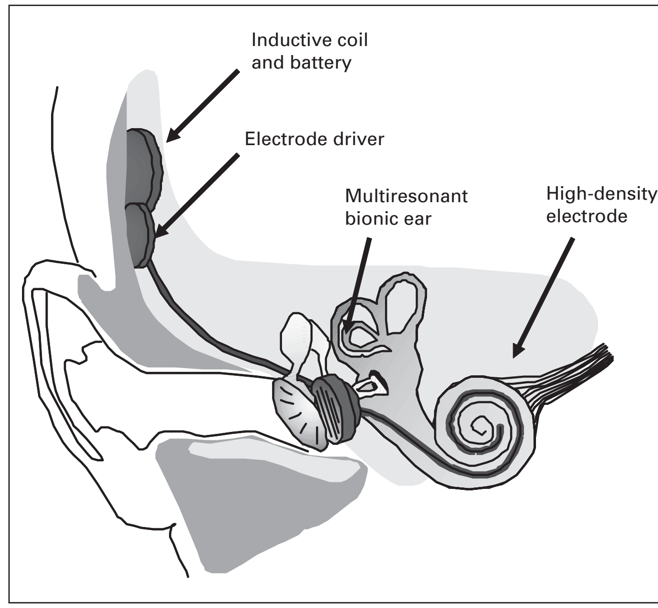

Fig. 1. Illustration of bionic ear concept. A multiresonant transducer receives acoustic energy and splits into frequency bands that mimic the tonotopic distribution of the cochlea. An electrode driver amplifies the signal and sends current to an implanted electrode in the cochlea.

An illustration of the optical microphone is shown in figure 1. It consisted of an array of suspended polymer cantilevers, each one at a different length, ranging from 2 to 7 mm. The cantilevers were rectangular cross section, 100 μm in width and 40 μm in height, made from epoxy using modern micromachining techniques for polymer [Xu et al., 2002]. The cantilevers were suspended over an etched cavity in silicon, allowing them freedom to vibrate. A second rectangular epoxy channel was fabricated to meet the cantilever at its distal end, stopping short of contact, leaving a 20-μm air gap. A 635-nm laser was directed down the cantilevers, and the light intensity was monitored at the exit end of the second epoxy channel. The transparent polymer channels acted as excellent light pipes, so that light was efficiently guided from the laser, through the channel and cantilever, through the second channel to the photodetectors at the end. When the cantilever vibrated, the cantilever was temporarily misaligned with its mating channel reducing the efficiency of light to pass across the 20-μm air gap. This was seen as a reduction in light intensity at the photodetector. In this

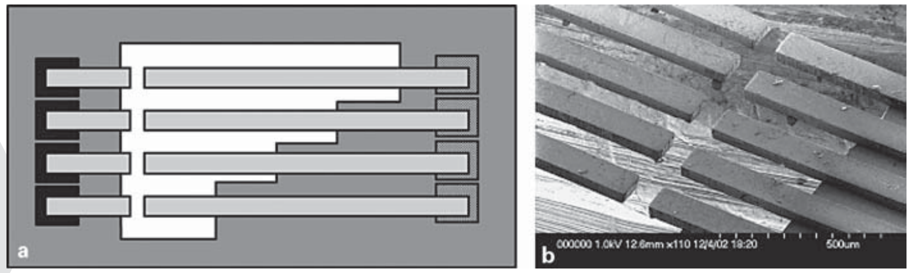

Fig. 2. **a** Illustration of four-channel multiresonant microphone showing cantilevers of different lengths suspended above an etched open cavity. **b** Scanning electron microscope image of cantilevers showing air gap between resonators and receiving light pipes.

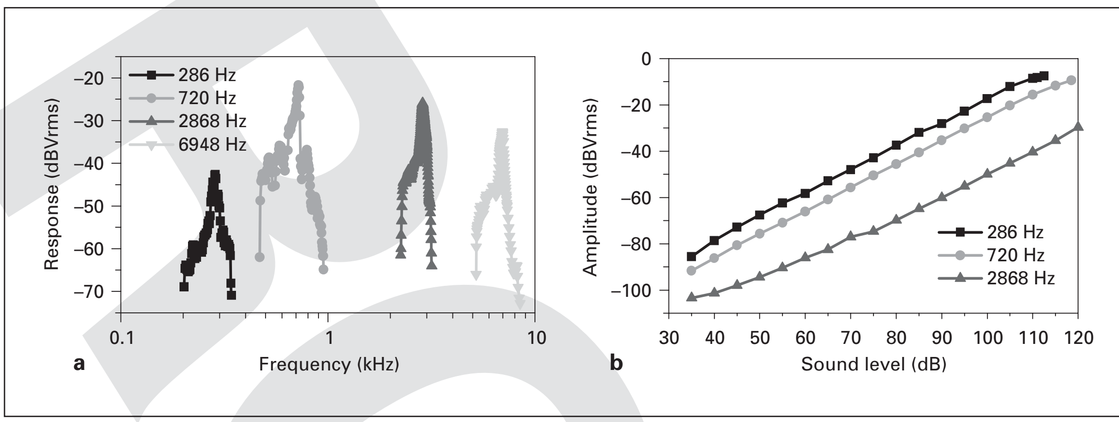

Fig. 3. Frequency response and dynamic response for multiresonant transducer. **a** The Q10 values of these cantilevers are 9.38, 10.11, 11.56, and 14.01 for resonant frequencies at 286, 720, 2868, and 6948 Hz, respectively. Measurements were made at 70 dB (SPL). **b** The polymeric cantilever array has a linear dynamic range of 80 dB for sound inputs between 35 and 115 dB (SPL). Measurements were made at each cantilever's resonance. The fourth cantilever is not shown in the right figure because it was destroyed during handling between measurements.

manner, the movement of the cantilever, and hence, the sound energy could be monitored (fig. 2).

Several variations of the optical device were built to demonstrate different fabrication methodologies. Fabrication methods included using UV patternable high-definition epoxy (SU-8), performing laser machining on polymer films, and performing microinjection molding. The details of the injection molding manufacturing process which produced the results presented here, have been detailed elsewhere [Xu et al., 2004].

We tested a four-resonator device by placing it under a speaker connected to an amplified tone generator. Signal was collected from the resonators and analyzed using standard data acquisition instrumentation. Frequency response, dynamic response, and directionality were recorded. The preliminary data, shown in figure 3, are very encouraging. Cantilever response shows specific peak frequencies at 286, 720, 2868, and 6948 Hz, respectively, well within human hearing range. Q10 values (peak frequency divided by the bandwidth 10 dB below the peak) are similar to mammalian basilar response [Robles and Ruggero, 2001]. Dynamic response is linear from 35 to 115 dB SPL. While linear response is an excellent characteristic for a microphone, for cochlear stimulation, dynamic compression may need to be performed using appropriate amplification circuitry.

We have observed similar results with cantilevers prepared for capacitive readout. In those devices, the cantilevers were coated with a thin (100-nm) layer of gold on their underside forming a capacitor between each cantilever and a ground plane directly beneath each cantilever. A bias of 45 V was placed on the cantilever making it a capacitor. Vibration of the cantilever resulted in changes in the capacitance, and thus modulated an induced current across the capacitor. The small signal was amplified by a JFET and recorded using conventional microphone

Table 1. Summary of differences between optical and electronic cantilevers

Optical readout	Electrical readout
Low noise floor	Noisy due to electromagnetic interference – good shielding required
Low bias	High bias preferred (45 V) or electret
Moderate power requirement due to light coupling losses (5–10 mW)	Low power (<1 mW)
Difficult integration with electronics	Easy integration with electronics
Good sensitivity and dynamic range	Good sensitivity and dynamic range

amplifiers and instrumentation. Ultimately, the capacitor (or even electret) readout is preferred over the optical readout because it is easier to integrate with conventional electronics and consumes considerably less power. However, the electrical system is more suspect to noise and must be carefully shielded, whereas the optical system demonstrated clear signal with almost no noise. Differences between electrical and optical readout are indicated in table 1.

The multiband transducer works because the individual cantilevers have been designed to exhibit resonances at frequencies within the range of human hearing. For a simple cantilever, the natural frequency is given by

$$f_k = \frac{1}{4\pi}\frac{n_k^2}{\sqrt{3}}\frac{T}{L^2}\sqrt{\frac{E}{\rho}}$$

where E = Young's modulus in pascals, T = thickness in meters, L = length in meters, ρ = density in kg/m^3, and n_k = 1.875, 4.694, 7.855, … (n_k is mode number). When energized by acoustic energy, the cantilever will respond with maximum amplitude at the natural frequency, as given by the well-known Lorentzian formula,

$$A(f) \propto \frac{\frac{\Gamma}{2}}{\left(f - f_0\right)^2 + \left(\frac{\Gamma}{2}\right)^2}$$

Here, Γ is the 'linewidth' or full width and half maximum. For discussion, we prefer to use the 'quality factor' value Q10, which is peak frequency divided by the bandwidth 10 dB below the peak, or Q10 = $f_0/3\Gamma$. Thus, high quality factors correspond to narrow resonances. The human cochlea is also a resonator and typically responds with Q10 values under 10, relatively low quality factors [Geisler, 1998]. Second and higher order modes will also be excited, but these are typically much lower in amplitude.

Traditional micromachining materials, namely silicon, ceramics and metals, characteristically exhibit large Young's modulus and low damping [Petersen, 1982]. This results in large natural frequencies (for example, a silicon cantilever, 1 mm × 5 μm, resonating at about 5 kHz) and high quality factors. These are desirable qualities for fabricating mechanical resonators, such as those used in miniature accelerometers and gyroscopes, but this is not satisfactory for mimicking the response of the cochlea. If the device can be built small enough, air may be used to dampen the oscillations [e.g., Haronian and MacDonald, 1995].

Polymers have more suitable material properties, exhibiting high damping and low modulus, typically 50 times less than metal. As a result, the natural frequencies of polymer cantilevers can be designed to be in the range of a few hundred Hz to 10 kHz for microphone size under 1 cm. Polymers have certain problems, however. Polymers cannot conduct electricity, requiring the addition of a thin metal layer if electrical transduction is desired. Polymers are difficult to fabricate at the small sizes required for this transducer. Polymers may exhibit long-term plastic deformation, or may develop stress from thermal processing. Indeed, our own microfabrication efforts required a special annealing step to reduce residual stress and straighten out the resonators (for microinjection molded cantilevers). Nevertheless, many engineered polymers exist that have been demonstrated as useful in critical applications, for example, polyester and polyimide.

Because natural frequency is so directly related to length (for a cantilever) it is easy to design multiband devices of arbitrary frequency distribution. Furthermore, since the signal from each cantilever is amplified, each channel's gain may be adjusted independently. In this manner, we can enable the design of a microphone with any arbitrary frequency range and response. We can imagine designing a transducer that can correctly compress and map electrical signals to all regions of the human cochlea.

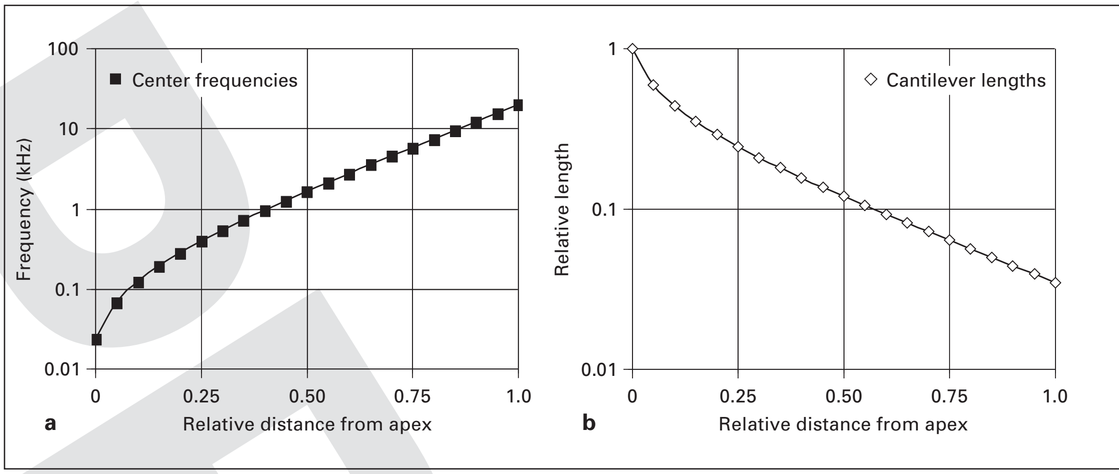

Fig. 4. a Tonotopic mapping of the human cochlea showing center frequency as a function of distance from the apex of the cochlea. **b** Relative lengths for a cantilever-based multiresonator that can reproduce natural frequencies that coincide with the center frequencies.

The mammalian cochlea has a tonotopic response to frequencies [Moore, 1997]. This relationship between center frequency and position along the basilar membrane has been mapped for several mammals and generally follows a relationship of

$$CF = A\,(10^{\alpha x} - k),$$

where CF is center frequency in kHz, x is the relative distance from the apex, $k \sim 0.85$ and $a \sim 1.2$ for most mammals [Greenwood, 1961, Greenwood, 1990, Robles and Ruggero, 2001]. The constant A determines the range of center frequencies (20 Hz–20 kHz in humans). This relationship indicates a logarithmic compression of frequencies at the high frequency range. A typical implantable electrode array is likely to produce electrodes at equally spaced separations, indicating that our resonator design should follow a similar compression in frequency. Cantilever resonators designed to mimic this frequency are readily fabricated using lithographic or UV cutting methods. Figure 4 shows the required cantilever lengths for electrodes destined to be placed at different regions in the cochlea. The cantilever length is given as a relative number since the physical length for a given center frequency also depends on the material and thickness of the cantilever, which can be adjusted according to design criteria.

Although this discussion assumes simple straight cantilevers, one is by no means limited to designing a system of uniform cantilevers only. One can achieve complex resonance profiles by assuming more complex shapes and mass distributions in the resonators. For example, resonators may employ torsional or meander springs, patterned mass areas, or material combinations in order to produce a desired response. For low-frequency response it would make more sense to increase the mass at the end of a cantilever rather than extend the length, enabling the transducer to remain small (fig. 5). In complex designs, the mechanical analysis is more sophisticated and finite element modeling is required. In many cases, bridge or ribbon structures may be preferred to cantilevers, particularly in the case of capacitive readout devices where

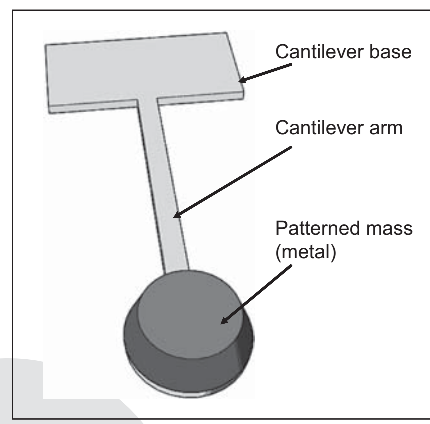

Fig. 5. Illustration showing method for decreasing natural frequency while still maintaining short cantilever length by adding additional mass at the end of the cantilever.

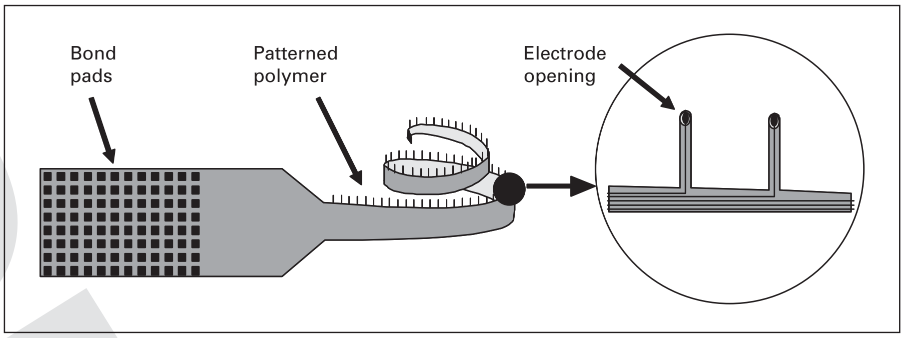

Fig. 6. Illustration of a high-density electrode concept. The polymer material and electrical traces may be completely defined by lithography, resulting in a large number of fine 'hairs' that contain electrodes. At the tip of each hair, the electrode is exposed allowing the electrode to penetrate close to the site of the hair cells, minimizing cross talk (and possibly threshold voltage) through the conductive cochlear fluid.

a small controlled gap between the resonator and the ground plane is required. A ribbon device can maintain tight gap tolerance, whereas any residual stress in a cantilever will result in bending, which will compromise the gap tolerance. Traveling wave phenomena may be mimicked by lightly coupling adjacent resonators through micromachined tethers or springs.

High-Density Microelectrode Arrays

The strategy of this and other technologies is to try to accurately mimic the response of the human cochlea so that one may artificially stimulate the cochlea in the way it was designed to be stimulated. It is unlikely, however, that true hearing can be restored unless the electrode density is made large. Small numbers of electrodes, blunt and ill-positioned, are likely to miscode and blend the spectral information of sound resulting in an unintelligible sensation. Electrode density is limited by practical concerns (manufacturability, power consumption), as well as by physical limitations – current lines tend to overlap for adjacent electrodes when the electrodes are far from their target, reducing the ability to stimulate specific sites. Thus, electrode design must also include a mechanism for the electrical contacts to be highly localized. The benefit of the micromechanical resonator is that a large number channels can be simultaneously filtered at low power and low latency, in a small package. Advanced, high-density electrodes are needed to complement this technology to deliver high-fidelity signals to the auditory nerves.

High-density electrodes may be manufactured using micromachining techniques similar to those used for building the resonator array. Figure 6 shows a hypothetical electrode array that can be manufactured in thin polymer membrane. The device consists of lithographically defined electrodes built up in platinum, passivated by ceramic or polymer (e.g., parylene), and encapsulated in a flexible polymer carrier, such as polyimide. Each electrode juts out in lithographically defined 'hair', 20–100 μm in width and several hundred micrometers in length. At the tip of each electrode hair is an opening in the passivation layer that exposes the platinum to the environment. Such hairs could enable the electrodes to make close contact with the basilar membrane and, presumably, minimize cross talk among nearby electrodes. One may wish to design multiple electrode hairs per electrical trace, and the hairs themselves may include hooks, dendrites, and other special geometries to improve electrical performance. Delamination of polymers may occur due to swelling from fluid exposure, failure of adhesives, or electrochemical effects such as cathodic delamination. As with any implantable device, materials reliability will be a critical factor for success.

Assuming the electrodes are 10 μm wide, with 10 μm interspacing (smaller electrical traces can be manufactured), one can trace out 50 electrodes in a single side of plastic, 1 mm in width. This suggests that an electrode capable of delivering all 88 keys on the piano should require a strip of plastic 1 mm in width, patterned on both sides with electrical traces. Since the electrode can be fabricated using conventional micromachining technology, one can imagine having each electrode strip custom produced to fit each patient's cochlea.

Smaller electrodes will result in higher resistances, increasing the driving power per electrode. The resistance of a 3-cm platinum electrode, 10×0.1 μm, will be nearly 3 kΩ. This will necessitate the use of smaller currents to reduce power consumption. Such a strategy can only work if smaller currents can still produce threshold voltages at the dendrites. One can anticipate that the hairlike electrodes indicated in figure 6 may experience lower thresholds because they are in such close proximity to the

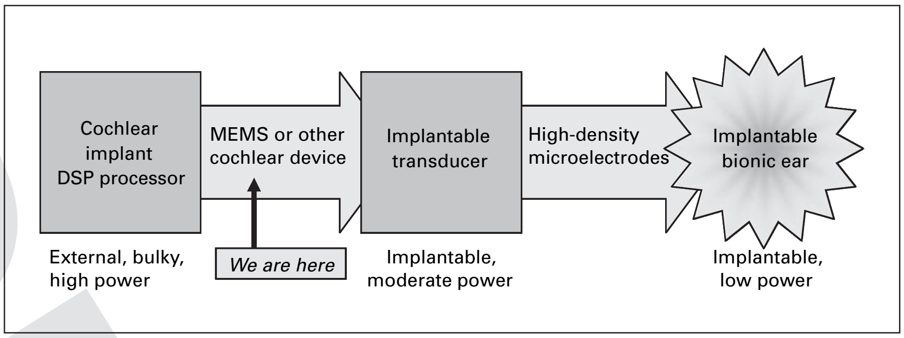

Fig. 7. Road map to bionic ear technology. Researchers are currently tackling the problem of building a miniaturized cochlear device. System insertion issues and, most importantly, high-density microelectrodes are critical developments for a successful bionic ear.

nerve sites, and less energy is wasted in the region between electrodes. This has not been experimentally verified, however, and more work is needed in this area to confirm design strategies.

System Packaging

A complete system can be expected to consist of a multiband microphone, amplification electronics, electrode driver, a high-density electrode array, small rechargeable battery, and a recharge coil. (The system might be copackaged with a traditional CI system as an optional secondary implant choice.) A significant engineering problem for such a system will consist of packaging for the microphone. There are several major issues that need to be addressed, namely (1) electrical packaging to make electrical connections from the microphone to the electronics, (2) mechanical packaging to mount the microphone in an appropriate location, (3) environmental packaging to seal and protect the microresonators from fluids, and (4) radio frequency packaging to shield the device from electrical noise.

The resonator array must be mounted so that a large number of resonators can make electrical contact to a microelectronic chip that performs the appropriate amplification for each channel. This may represent a large number of bond points, possibly hundreds of electrical connections may need to be made. High density bump-bonding, or even postprocessing of the microfabrication directly on the die are possible solutions to this problem. Since the fabrication method can be designed to be performed at low temperature, one may consider building the microresonators directly on the electronic die.

Direct connection of the resonators to the mechanical substrate can degrade performance of the microphone. Vibrations of the mechanical package can be readily picked up by the transducer, typically introducing broadband response where narrow band may be desired. This is a well-known problem for microphone designers. One may need to design damping systems or a vibration isolation mechanism into the packaging or into the microdevice itself.

Sealing the device against fluid leakage is a particularly difficult task because the protective package will introduce an acoustic barrier and impedance mismatch which will degrade the performance of the transducer. One approach is to follow the example of the reptilian middle ear and use a columella (a stiff rod) to connect the ear drum to a membrane opening (analogous to the oval window) in the packaged device. By choosing the size of the window appropriately, one may be able to match the acoustic impedances.

For most electrical transducers (e.g., capacitive, magnetic), interference from external electromagnetic sources is very problematic and greatly increases the noise in the signal. All condenser and electret microphones are heavily shielded against electromagnetic interference through metal packages and grills. One may hope that the presence of conductive fluid in the ear chamber and head can help provide natural shielding for the microphone. If not, then conductive casing will need to be placed around the transducer, grounding the system to the electrical potential of the patient.

Summary

We describe a micromachined multiresonator technology for building an artificial human cochlea that allows flexible design and good integration with electronic circuitry. The use of polymer material is recommended for low Q characteristics. An array of resonating cantilevers, each built with a different natural frequency, allows a device to perform a mechanical Fourier transform at

the front end of a bionic ear system. The channels may be mechanically coupled together, if desired. Furthermore, by controlling the amplification gain and the composition and geometry of the resonators, one may achieve sophisticated frequency profiles for each sub-band channel. The sub-band signals can be used to directly stimulate the cochlea according to its tonotopic arrangement. A mechanical bank of resonators can only be considered for this application if the resonators are very small, so that the device can be implanted in the ear cavity of a patient. Miniaturization methods, developed for electronic and sensor applications, can now be directed to make such small resonators.

A number of technologies are being explored by researchers to build artificial human cochleas, ranging from microfluidic devices, micromechanical devices, and electronic devices. A possible roadmap to a bionic ear is shown in figure 7. A miniaturized cochlear device is not enough, however. A critical development for the implant to be useful is the technology to build high-density electrode arrays that can efficiently bring the many sub-band signals to the appropriate nerve endings. System engineering issues, such as electronic integration, power sources, and sophisticated packaging also need to be studied and understood.

Ultimately, the goal for this type of technology is to simulate the response of the human cochlea. Any analog approach, whether fluidics, mechanics or analog electronics will lack the flexibility of digital programming. Analog strategies are likely to be most successful when combined with digital control electronics to provide a measure of programmability for each individual patient.

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