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# Polymeric micro-cantilever array for auditory front-end processing

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**Abstract**

A polymeric micro-cantilever array has been developed that mimics the biological front-end processing in the mammalian cochlea, and is intended for use in applications of auditory prostheses. Made of optical epoxy polymer molded over a silicon substrate, the micro-cantilevers have similar mechanical performance as the basilar membranes in the mammalian ears. The polymeric cantilevers, which are transparent, are used as optical waveguides to guide and modulate a light beam, which is initiated from a light emitting diode (LED) and collected by a photo diode, to produce a signal suitable for human hearing when the sound wave excites them. The polymeric cantilevers have $Q_{10}$ values of 9.38, 10.11, 11.56, and 14.01 for resonant frequencies at 286, 720, 2868, and 6948 Hz, respectively. These values are similar to those obtained by direct measurement of the basilar membrane. Furthermore, they have a linear dynamic range of more than 80 dB sound pressure level (SPL) with less than 15% total harmonic distortion (THD). This polymeric micro-cantilever array has low power consumption, short processing time, high sensitivity, high frequency resolution, small size, is insensitive to electromagnetic interference, and is suitable for a totally implantable device in the human ear.


*Keywords:* Cantilever array; Polymeric cantilever; Acoustic sensor; Auditory prostheses; Mechanical filter; Cochlear implant; Hearing aid

## 1. Introduction

Ten percent of the general population suffers from hearing loss and that number rises to 35% for people over 65 years old. People who are 25–45-year-old make up 42% of the hearing-impaired population while people who are older than 65 make up 45%. Nowadays, new hearing devices such as digital hearing aids and cochlear implants have significantly improved the quality of life for hearing-impaired people [1–3]. A microphone and signal processor are two critical functional components in these devices. The microphone senses sound and the processor analyzes it. Currently digital signal processing (DSP) is usually used to process the sound for its flexibility and programmability [4,5]. However, apparent drawbacks such as relatively large size, high power consumption and long processing time associated with a large number of channels in the digital signal processing (DSP) technique increase its cost and size and limit its utility. Different from these hearing devices, the mammalian cochlea performs sensing and analyzing of sound by the basilar membrane, a single component simultaneously. With the same function as basilar membrane, micro-mechanical filters are presently an alternative to solve these issues that appear in the current hearing devices [6,7]. Using fabrication technology that was borrowed from traditional semiconductor processes, present cantilevers in micro-mechanical filters are made of high Young's modulus materials, such as silicon [8] and silicon nitride [9], and possess a high quality factor. It is known that a higher quality factor means more cantilevers are required to cover the whole audio frequency range. That will increase the power consumption, complicate the device fabrication, and enlarge the device size, as well as the complex control and processing circuits. However, physiological measures have shown that the basilar membranes in the mammalian cochlea, which are acoustic sensors in the mammalian ears, have relatively low quality factor $Q_{10}$ ($Q_{10}$ is the peak frequency divided by the bandwidth 10 dB below the peak.) between 1 and 10 for resonant frequencies in the audio frequency range [10]. These $Q_{10}$ values are two orders of magnitude lower than that of the present mechanical filters.

Here we report a polymeric micro-cantilever array that mimics the biological front-end processing in the cochlea and can be applied to both advanced hearing aids and cochlear implants. We use an optical polymer, a material of low Young's modulus, to produce similar mechanical

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performance to the natural organic material found in the basilar membranes. We also take advantage of the polymeric cantilever's transparence property to use it as an optical waveguide to guide and acoustically modulate a light signal. The polymeric micro-cantilever array has low power consumption due to passive operation, short processing time due to parallel mechanical filtering, few channel requirement due to low quality factor, is insensitive to electromagnetic interference and has high sensitivity due to optical detection. In addition, its high frequency resolution and small size as a result of microfabrication make it suitable for a totally implantable device in the human ear. No body-worn devices will be needed, unlike those required for the hearing devices based on DSP.

## 2. Design

The polymeric micro-cantilever array, shown in Fig. 1, consists of mechanical substrate, a light emitting diode (LED) array, a freestanding polymeric cantilever array, and a photo diode (PD) array. Light originates from the LED and is directed through the polymeric cantilever, which is also an optical waveguide due to the polymer's transparence and higher refractive index than air, to the PD. When the sound wave reaches the cantilever, the cantilever resonates if the sound wave frequency matches its resonant frequency. As a result, the light intensity detected by PD is modulated by the cantilever only for resonant frequencies. By combining an array of cantilevers that have different lengths and so different resonant frequencies, we can accomplish both sensing and frequency filtering functions at the PD array output. In other words, a sound spectrum can be obtained at the PD array output with the cantilever array.

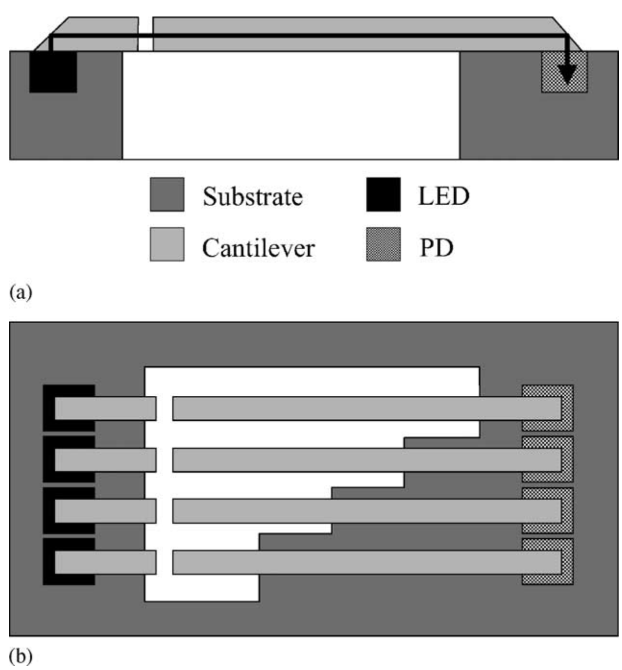



Fig. 1. Schematic of the polymeric micro-cantilever array. (a) Cross-section and (b) top view. Light originates from the light emitting diode (LED) and directed through the polymeric cantilever to the photo diode (PD). When the sound wave reaches the cantilever, the cantilever resonates if the sound wave frequency matches its resonant frequency. As a result, the light intensity detected by photo diode is modulated by the cantilever. By combining an array of cantilevers with different resonant frequencies, the sound spectrum can be obtained.

For our experiments, we used a polymer of Epo-tek UVO-114 optical epoxy and designed cantilevers with thickness of 50 μm, width of 100 μm, and lengths of 1.5, 2.4, 4.8, and 7.5 mm, producing resonant frequencies of 7374, 2880, 720, and 295 Hz, respectively.

## 3. Fabrication

In conventional technology, polymeric waveguides and cantilevers are fabricated by photolithography and reactive ion etching (RIE). However, etching polymer is very slow and difficult. To solve this problem, we developed a vacuum injection molding process to build our polymeric cantilevers. This process is simple, low-cost, and scalable to mass production [11]. With this process, the polymer is injected into a micro-mold of inert silicone elastomer, poly(dimethyl)siloxane (PDMS). The whole process consists of the following five steps.

### 3.1. Mold preparation

A high resolution photopatternable epoxy, SU-8 photoresist, was spun on a silicon substrate to 50 μm thickness, shown in Fig. 2(a). After prebake was performed at 65 °C for 6 min then 95 °C for 20 min, photolithography was employed to transfer a cantilever pattern from a chrome mask to the SU-8 photoresist layer. After 90 s exposure, post-exposure baking was performed at 65 °C for 1 min then 95 °C for 5 min. Following about 8 min development, a hard bake was performed at 150 °C for 15 min to obtain good mechanical strength for the SU-8 material, shown in Fig. 2(b).

Uncured poly(dimethyl)siloxane elastomer (PDMS) was cast over the SU-8 substrate to about 2 mm thickness, and then cured in oven at 90 °C for 15 min, shown in Fig. 2(c). After curing, the PDMS layer was gently peeled off from the substrate. The PDMS material has excellent release properties, and separates from micromachined surfaces without the need for release agents. The 2 mm thick PDMS mold contained the cantilever pattern for the device, shown in Fig. 2(d). Small holes were punched in the mold for subsequent polymer loading in the vacuum injection molding process, shown in Fig. 2(e). These small holes were only used for polymer loading and would be cut out at the final dicing process, so they did not need to be made too precisely.

### 3.2. Substrate preparation

An optical fiber was used to couple visible light in to and collect the light out from the cantilevers. For future studies,

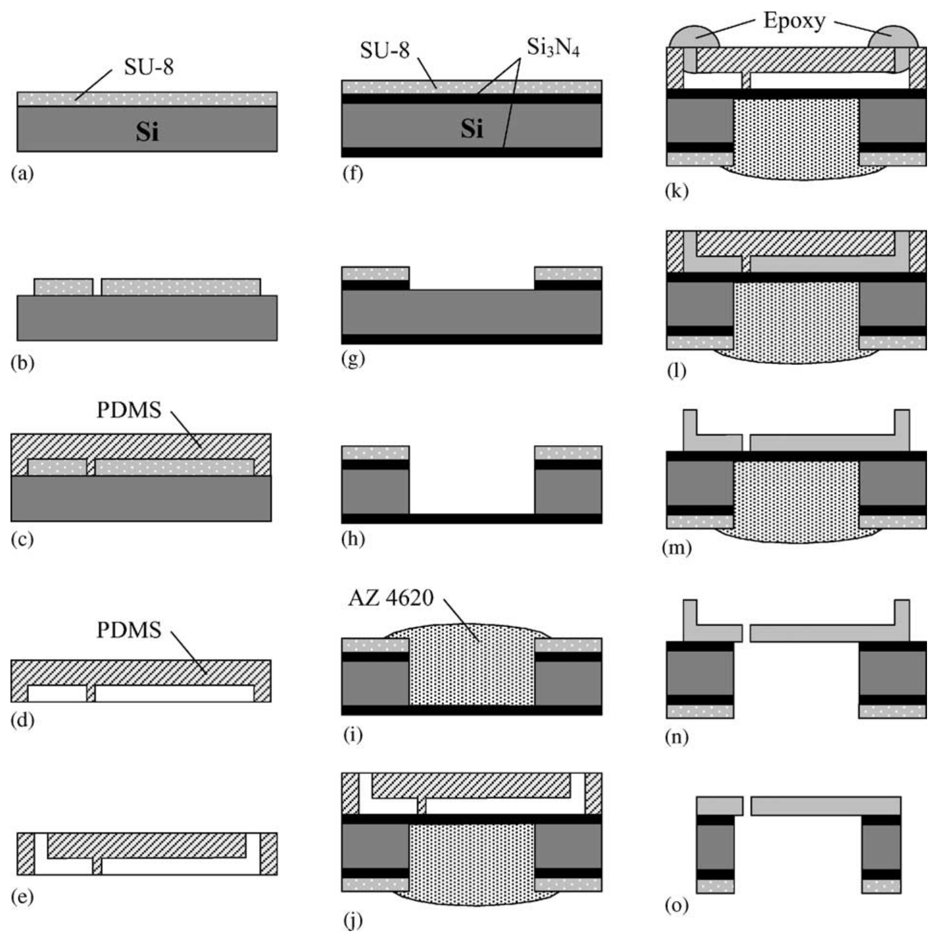



Fig. 2. Process flow of the fabrication of polymeric micro-cantilever array. The mold preparation is shown from (a) to (e), the substrate preparation from (f) to (i), the vacuum injection molding from (j) to (m), the cantilever release in (n), and the annealing and polish in (o).

the substrate will include an integrated LED array and PD array.

We started the process with a silicon wafer (P-type, 525 μm thick) that had a 3000 Å LPCVD $Si_3N_4$ layer. SU-8 photoresist was spun on it to ∼50 μm thickness, shown in Fig. 2(f). After the SU-8 photoresist was patterned by photolithography, the $Si_3N_4$ was removed by reactive ion etch (RIE) using gases of $CF_4$ and $O_2$ to open the Si etching window, shown in Fig. 2(g). The Si in the etching window was etched by deep RIE (using gases of $SF_6$, $O_2$, and $C_4F_8$) and leaving the $Si_3N_4$ membrane on the other side, shown in Fig. 2(h). AZ 4620 photoresist was dropped in the well and cured to form a sacrificial material, shown in Fig. 2(i). This photoresist sacrificial layer is critical to protect the polymeric cantilevers and the $Si_3N_4$ membrane when the PDMS mold is peeled off in the vacuum injection molding process step. The subsequent process was performed on the other side.

### 3.3. Vacuum injection molding

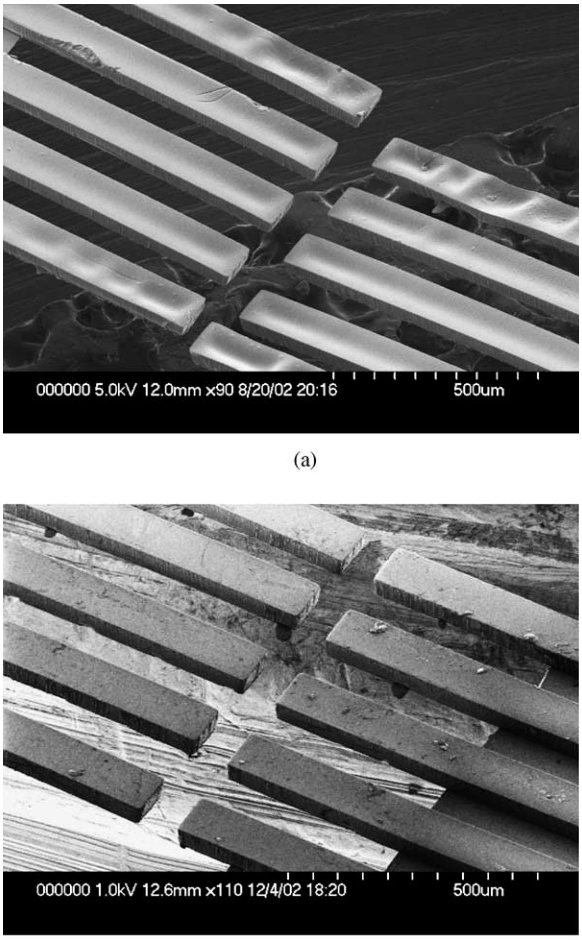
The PDMS mold was positioned on the substrate and pressed softly. The soft silicone material adhered to the $Si_3N_4$ surface, shown in Fig. 2(j). The assembly was placed in a vacuum chamber and vacuum was turned on for 15 min, then uncured Epo-tek UVO-114 optical epoxy was dropped in the loading holes by use of custom tooling, shown in Fig. 2(k). With our special tool, we can drop the epoxy under good control. Next, the chamber was allowed to return to atmospheric pressure. The liquid polymer was injected into the channels by the atmospheric pressure, shown in Fig. 2(l). After curing the polymer, the mold was easily peeled from the top due to the PDMS's excellent release property, shown in Fig. 2(m).

### 3.4. Cantilever release

The AZ 4620 photoresist was dissolved in acetone and the $Si_3N_4$ was etched from the back side by RIE (using gases of $CF_4$ and $O_2$) to release the cantilevers, shown in Fig. 2(n).

### 3.5. Annealing and polish

After release, the polymeric cantilevers usually bent due to internal stress accrued during processing, shown in Fig. 3(a).




(a)

(b)

Fig. 3. SEM micrograph of the polymeric micro-cantilever array (a) before annealing and (b) after annealing. An annealing at 90 °C for 72 h and return to room temperature slowly over a 3 h period can make the polymeric cantilevers straight. A good alignment of cantilevers ensures a high coupling efficiency of light.

To solve this problem, an annealing process was employed. The cantilevers were annealed at 90 °C for 72 h, then allowed to return to room temperature slowly over a 3 h period with temperature ramp of 10 °C reduction in 30 min. After annealing, the polymer cantilevers straightened and aligned very well with each other, shown in Fig. 3(b). Finally, the device was diced and polished at the ends to facilitate good light coupling between the optical fibers and the waveguides, shown in Fig. 2(o). Fig. 4 is a SEM micrograph that shows the 20 μm gap between the cantilevers. The smooth and even gap ensures a good light coupling between the polymeric cantilevers.

## 4. Measurement

The testing system consisted of a laser diode driver, a pigtailed laser diode, input and output optical fiber, the polymeric cantilever array, optical power meter, amplifier,

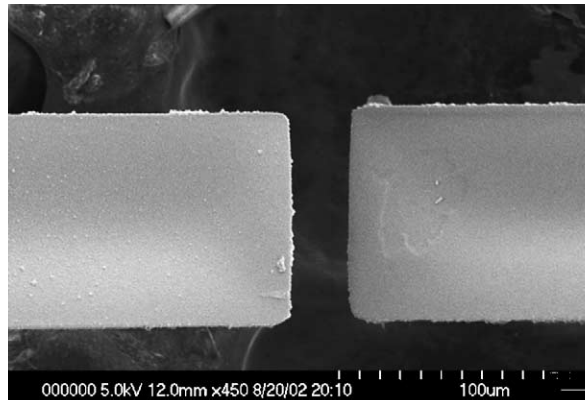



Fig. 4. SEM micrograph of the 20 μm gap between the polymeric cantilevers. The smooth and even gap ensures a good light coupling between the polymeric cantilevers.

dynamic signal analyzer, speaker, and sound level meter. The schematic of setup is shown in Fig. 5.

The light originated from the laser diode and passed through the input fiber, polymeric cantilever, and output fiber to the optical detector. Fig. 6 shows the light went through the left polymeric cantilever, crossed the 20 μm gap, and coupled into the right polymeric cantilever. In this picture, the light was only launched into the middle cantilever. A dynamic signal analyzer generated an electrical signal to drive the speaker to produce sound. The sound excited the cantilevers, making them resonate when sound frequency matched their resonant frequency, and consequently modulating the coupling efficiency of the light. The modulated optical signal was converted to electrical signal by the optical detector. Then the electrical signal was amplified and sent back to the dynamic signal analyzer for frequency analysis. The sound level meter was placed near the cantilevers to measure the sound level. The characteristics of polymeric cantilevers, such as frequency response, dynamic range, total harmonic distortion (THD), cross-talk between cantilevers, and group delay, were all measured in the experiments.

## 5. Results and discussion

### 5.1. Frequency response

Fig. 7 shows the frequency response of the polymeric cantilever array. These cantilevers have resonant frequencies of 286, 720, 2868, and 6948 Hz. These frequencies are little lower than design values because the actual thickness of cantilevers was little thinner than design. The $Q_{10}$ values of cantilevers are 9.38, 10.11, 11.56, and 14.01 for resonant frequencies at 286, 720, 2868, and 6948 Hz, respectively. They are the same order of magnitude as, but little higher than, those that obtained with direct measurement of the basilar membrane vibration in a normal mammalian cochlea. These direct measurement values are between 1 and 10 for resonant

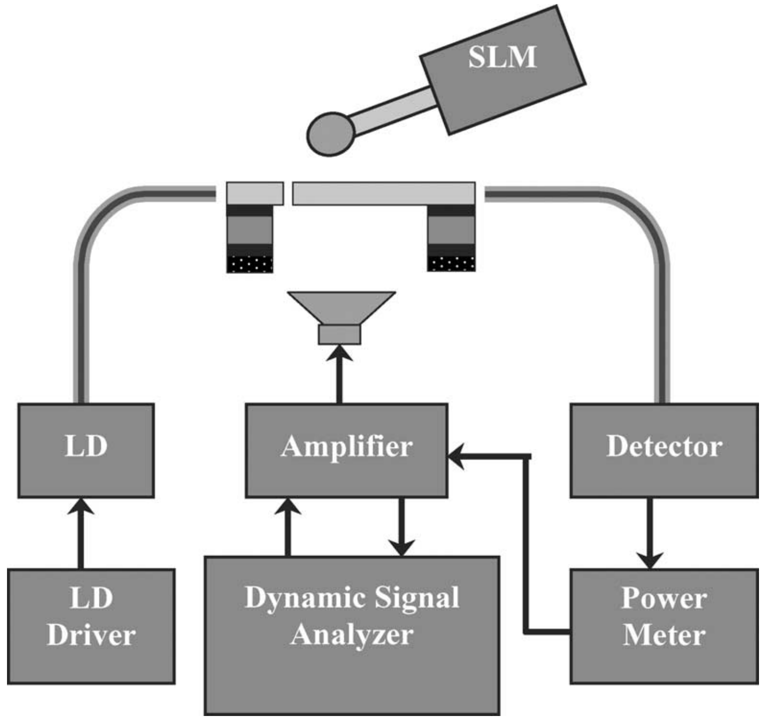



Fig. 5. The schematic of testing system for cantilever characterization. The light originated from the laser diode (LD) and passed through the input fiber, polymeric cantilever, and output fiber to the optical detector. A dynamic signal analyzer generated an electrical signal to drive the speaker to produce sound. The sound resonated the cantilevers, modulating the coupling efficiency of the light. The modulated optical signal was converted to electrical signal, and then amplified and sent back to the dynamic signal analyzer for frequency analysis. A sound level meter (SLM) was placed near the cantilevers to measure the sound level.

frequencies in the audio frequency range [10]. Lower $Q_{10}$ values can be obtained by choosing a polymer with lower Young's modulus in future designs.

The low $Q_{10}$ value is important for the application of polymeric cantilevers to hearing devices. A lower quality factor means larger bandwidth, thus, fewer cantilevers are required to cover the whole audio frequency range. In addition, lower quality factor means higher damping factor, thus, less interference from ambient vibration, and no "ringing" after a sound has been heard.

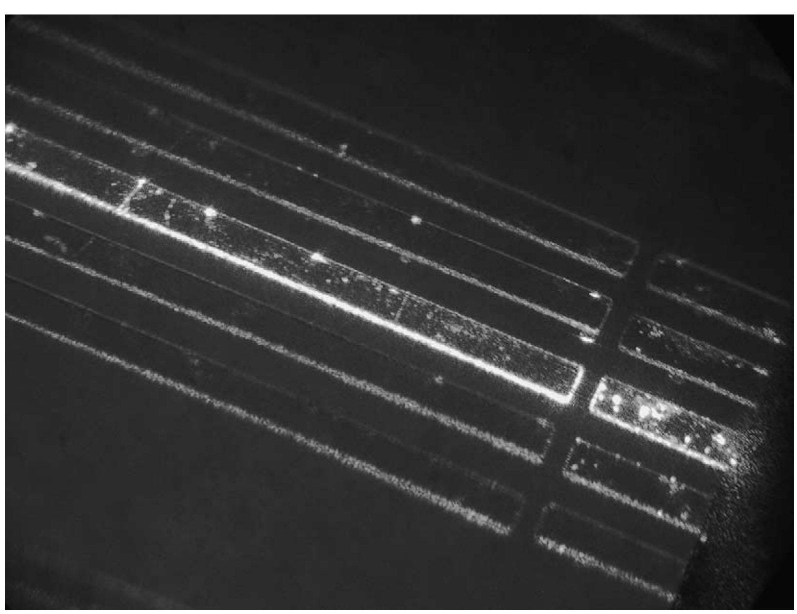


Fig. 6. Optical photograph that shows the light passing through the left polymeric cantilever, crossing the 20 μm gap, and coupling into the right polymeric cantilever. The light was only launched into the middle cantilever.

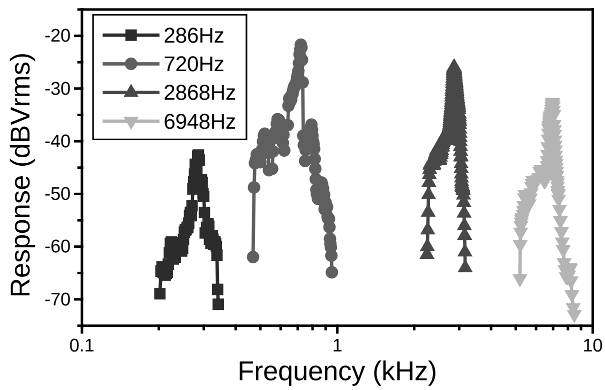



Fig. 7. The frequency response curves of the cantilever array. The $Q_{10}$ values of these cantilevers are 9.38, 10.11, 11.56, and 14.01 for resonant frequencies at 286, 720, 2868, and 6948 Hz, respectively. They are the same order of magnitude as, but little higher than, those that obtained with direct measurement of the basilar membrane vibration in a normal mammalian cochlea. Lower $Q_{10}$ values can be obtained by choosing a polymer with lower Young's modulus in future designs.

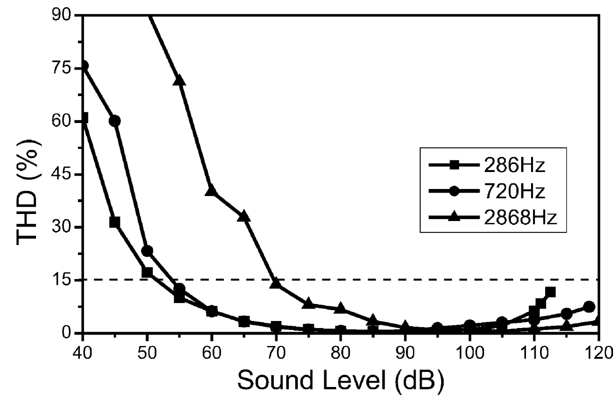



Fig. 9. The total harmonic distortions (THD) of the polymeric cantilever array. The THD is generally low for loud sounds, but appears to be higher than 15% when the sound level is lower than 50, 55, and 70 dB for the frequencies at 286, 720, and 2868 Hz, respectively. We believe the high distortion reflects the existence of ambient background vibrations, which interfered with the signal and the dynamic signal analyzer interpreted the noise as a harmonic component to calculate the THD.

### 5.2. Dynamic range

Fig. 8 shows the input/output function of the polymeric cantilevers responses to tones. It is seen that they have a linear dynamic range for sound inputs between 35 and 115 dB SPL. The lower limit of 35 dB is due to background noise while the upper limit of 115 dB is due to the power of amplifier in our measurement. Therefore, the polymeric cantilever array should have a linear dynamic range of more than 80 dB SPL.

### 5.3. Total harmonic distortion

The total harmonic distortions of the polymeric cantilever array are shown in Fig. 9. This figure shows the harmonic distortion is generally low for loud sounds, but appears to be higher than 15% when the sound level is lower than 50, 55, and 70 dB for the frequencies at 286, 720, and 2868 Hz, respectively. We believe the high distortion reflects the existence of ambient background vibrations, which interfered with the signal and the dynamic signal analyzer interpreted the noise as a harmonic component to calculate the THD. Proper acoustic packaging will help isolate ambient vibrations from airborne sound, and can reduce the THD at low sound intensities, and we are currently designing such protective housing for the sensor.

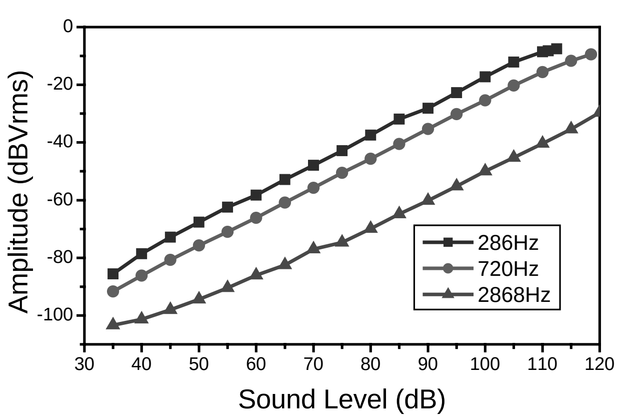



Fig. 8. The input/output functions of the polymeric cantilever array responses to tones. The polymeric cantilever array has a linear dynamic range of 80 dB for sound inputs between 35 and 115 dB SPL. The lower limit of 35 dB is due to background noise while the upper limit of 115 dB is due to the power of amplifier in our measurement.

### 5.4. Cross-talk

The cross-talk between cantilevers is an important parameter for micro-cantilever array that is in micrometer scale. In our device, the distance between adjacent cantilevers is 100 μm. Light was launched into a cantilever and the light out of this cantilever and two adjacent cantilevers was measured. The cross-talk between cantilevers can be obtained by comparison of the output optical power from these three cantilevers. The results show the cross-talk between cantilevers is smaller than −20 dB, which satisfies the typical requirements for hearing applications.

### 5.5. Group delay

Group delay indicates the processing speed of the micro-cantilever array signal. The results show the polymeric micro-cantilever array has a group delay of about 9 ms.

## 6. Conclusions

A prototype of the polymeric micro-cantilever array has been designed, fabricated, and measured. These polymeric cantilevers have the $Q_{10}$ values of 9.38, 10.11, 11.56, and 14.01 for resonant frequencies at 286, 720, 2868, and

6948 Hz, respectively. They have a linear dynamic range of more than 80 dB from 35 to 115 dB SPL and less than 15% harmonic distortions throughout most of the dynamic range of human hearing. The encouraging preliminary data demonstrate that the polymeric micro-cantilever array has similar mechanical performance to the basilar membranes in the cochlea and represents a feasible alternative to the digital signal processing technology in hearing devices.

The polymeric micro-cantilever acoustic sensor array has low power consumption, short processing time, high sensitivity, and is free of electromagnetic interference. In addition, its high frequency resolution and small size as a result of microfabrication make it suitable for a totally implantable device in the human ear.

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## References

[1] K.J. Dormer, M.A. Phillips, Auditory prostheses: implantable and vibrotactile devices, IEEE Eng. Med. Biol. Mag. 6 (1987) 36–41.

[2] S.U. Ay, F.G. Zeng, B.J. Sheu, Hearing with bionic ears, IEEE Circuits Devices Mag. 13 (1997) 18–23.

[3] K.D. Wise, K. Najafi, Fully-implantable auditory prostheses: restoring hearing to the profoundly deaf, Tech. Digest, International Electron Devices Meeting, 2002, pp. 499–502.

[4] H. McDermott, A programmable sound processor for advanced hearing aid research, IEEE Trans. Rehabil. Eng. 6 (1998) 53–59.

[5] P.C. Loizou, Signal processing for cochlear prosthesis: a tutorial review, in: Proceedings of Midwest Symposium on Circuits and Systems (MWSCAS'97), Sacramento, CA, USA, August 1997, pp. 200–204.

[6] M. Harada, N. Ikeuchi, S. Fukui, S. Ando, Fish-bone-structured acoustic sensor toward silicon cochlear systems, vol. 3514, in: Proceedings of the SPIE, Santa Clara, CA, USA, September 1998, pp. 266–275.

[7] M. Muller, H. Toshiyoshi, H. Fujita, Acoustic wavelet analysis using micro electro-mechanical sensors, vol. 3514, in: Proceedings of the SPIE, Santa Clara, CA, USA, September 1998, pp. 322–330.

[8] M. Harada, N. Ikeuchi, S. Fukui, H. Toshiyoshi, H. Fujita, S. Ando, Micro mechanical acoustic sensor toward artificial basilar membrane modeling, Trans. IEE Japan 119 (1999) 125–130.

[9] S.H. Shen, S.T. Yong, W. Fang, Design and fabrication of a MEMS filter bank for hearing aid applications, in: Proceedings of the 2nd Annual International IEEE-EMBS Special Topic Conference on Microtechnologies in Medicine & Biology, Madison, WI, USA, May 2002, pp. 352–355.

[10] L. Robles, M.A. Ruggero, Mechanics of the mammalian cochlea, Physiol. Rev. 81 (2001) 1305–1352.

[11] T. Xu, Z. Lai, Y. Yang, M. Bachman, G.P. Li, A low-cost injection-molded polymeric channel waveguide, in: Proceedings of the Optical Fiber Communication Conference (OFC) 2003, Atlanta, GA, USA, March 2002, pp. 321–323.

## Biographies

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*Mark Bachman* received his PhD in experimental physics from the University of Texas, Austin, in 1993. He is currently adjunct assistant professor of electrical engineering and computer science at UC-Irvine, an affiliate faculty in UCI's Department of Biomedical Engineering, and Assistant Director of UCI's Integrated Nanosystems Research Facility. Professor Bachman maintains an active research program in sensors, MEMS, BioMEMS, and micro-instruments for applications in communications and life sciences. He has lead microsystems-based research projects for DARPA, NIH, NSF and UC Discovery, as well as for major contracts from industry affiliates. Professor Bachman's research interests address the development of new methods of integrated microsystems engineering for applications beyond microelectronics. His work has led to the development of new micromachining and integration techniques for metals and polymers, integration methods for optics and fluidics, and a variety of microdevices and microsystems with applications in wireless, optical telecom, fluidics, medicine and biotechnology.

*Fan-Gang Zeng* received his bachelor degree in electrical engineering from the University of Science and Technology of China in 1982, an MS in biomedical engineering from Shanghai Institute of Physiology, Academia Sinica in 1985, and a PhD in hearing and neural sciences from Syracuse University, New York in 1990. He is currently Research Director in the Department of Otolaryngology—Head and Neck Surgery, and Director of the Hearing and Speech Research Laboratory at the University of California, Irvine. He has more than 50 publications in these areas including two papers in science magazine and three patents in the design of cochlear implants. He has given more than 110 presentations at national and international professional meetings, including 60 invited or keynote talks. He is currently a member of the NIH Integrative, Cognitive, and Functional Neuroscience Study Section 6 and associate editor of *IEEE Transaction on Biomedical Engineering*.

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