Development of Neural Probe with Microfluidic Channel Fabricated by Using Wafer Direct Bonding Technique

Soichiro Kanno\textsuperscript{1}, Risato Kobayashi\textsuperscript{2}, Takafumi Fukushima\textsuperscript{2}, Kazuhiro Sakamoto\textsuperscript{3}, Norihiro Katayama\textsuperscript{4}, Hajime Mushiak\textsuperscript{2}, Tetsu Tanaka\textsuperscript{1,2}, and Mitsumasa Koyanagi\textsuperscript{2},

\textsuperscript{1} Department of Biomedical Engineering, Graduate School of Biomedical Engineering, Tohoku University
\textsuperscript{2} Department of Bioengineering and Robotics, Graduate School of Engineering, Tohoku University
\textsuperscript{3} Research Institute of Electrical Communication, Tohoku University
\textsuperscript{4} Department of Applied Information Science, Graduate School of Information Sciences, Tohoku University
\textsuperscript{5} Department of Physiology, Tohoku University School of Medicine

1. Introduction

Drug Delivery System (DDS) is the system which controls drug distributions in a body quantitatively, spatially, and temporally. The DDS will make it possible to increase drug effect while reducing drug side effect. By adding the drug delivery functions to neural probe, we can record neuronal signals when drugs are injected around neurons [1]. Moreover, this kind of neural probes will be also useful in treatment of brain diseases by transporting drugs into the brain depths. These functions can be realized by integrating microfluidic channels with the neural probe.

As we have been developing the brain signal processing system having the neural probe, it is valuable to include the drug delivery function in the system [2]. Figure 1 shows the configuration of the brain signal processing system having the neural probe array with microfluidic channel, micro pumps, signal processing circuits, flexible cables with secondary coil for inductive link, main control circuits, and various kinds of sensors. Micro pumps and signal processing circuits such as amplifiers, multiplexers, and A/D converters are directly mounted on the neural probe array by using multi-chip bonding techniques. We can record different kinds of data such as electrical, chemical, and optical data simultaneously with injecting drug solutions.

In this paper, we designed and fabricated a Si micro needle with microfluidic channel for neural probing. Furthermore, we investigated fluidic characteristics of microfluidic channel in order to administer fluidic medicine.

2. Fabrication of Si micro needle with microfluidic channel

Figure 2 illustrates a micro needle with microfluidic channel we designed. The microfluidic channel was formed by using wafer direct bonding techniques. Two wafers are bonded by not an adhesive but covalent bonding of atoms [3]. The wafer direct bonding is formed by hydrogen bonds between oxidized wafer surfaces. The oxidized surface is formed by RCA1 rinse for one wafer, and is formed by thermal oxidation for the other wafer. When we stack two wafers, hydrogen bonds are formed.

By annealing stacked wafers, hydrogen bonds changed to covalent Si-O bonds accordingly with the reaction below: $\text{SiOH} - \text{OH}_2 - \text{OH}_2 - \text{OHSi} \leftrightarrow \text{Si} - \text{O} - \text{Si} + 3\text{H}_2\text{O}$. (1)

Figure 3 shows the overall structure of the fabricated needle. This needle has the length of 40 mm, the tip width of 100 $\mu$m, and the thickness of 280 $\mu$m. In addition, the microfluidic channel has the length of 40 mm, the width of 30 $\mu$m, and the depth of 10 $\mu$m. The fluidic outlet is $10 \times 10 \mu$m$^2$ large, and the fluidic inlet is $30 \times 10 \mu$m$^2$ large. The cross-sectional SEM image of the fabricated microfluidic channel is shown in Fig. 4. It is confirmed that two wafers were completely bonded.

3. Results and Discussion

The fluidic experimental system consists of a syringe pump and a pressure meter. At first, we confirmed the strength of the fluidic channel. In the experiment, we injected red ink into the fluidic channel with flow pressure of 1000 mmHg. Figure 5 shows the ejection of red ink from the fluidic outlet. From this experiment, it is confirmed that the microfluidic channel has enough strength to inject fluidic medicine.

We also evaluated relationships between the flow rate and the pressure drop through microfluidic channel. We changed a flow rate from 0.1 to 0.4 $\mu$m/min with the syringe pump, and we measured the pressure drop across the channel by the pressure meter. We recorded the fluidic characteristics of microfluidic channel, and compared with theoretical calculations. The pressure drop ($\Delta p$) is defined as:

$$\Delta p = \frac{\mu V (Re \times f) L}{2D^2}$$  \hspace{1cm} (2)

where $\mu$ is the viscosity of fluid ($\sim 8.9 \times 10^{-4}$ kg/ms at 25 $^\circ$C for water), $Re \times f$ is the product of the Reynolds number and the Darcy friction factor ($\sim 64$ in laminar flow) [4], L is the length of the channel (40 mm), and D is the hydraulic diameter of the channel as defined by:

$$D = 4 \times \left( \frac{\text{Cross-sectional area}}{\text{wetted perimeter}} \right)$$  \hspace{1cm} (3)
As shown in Fig. 6, the measured values agreed well with the theoretical calculations.

4. Conclusion

We successfully fabricated the Si micro needle with microfluidic channel for neural probing by using wafer direct bonding techniques. From fluidic experiments, it becomes clear that the bonding strength of the bonded wafers withstand the flow pressure larger than 1000 mmHg. Additionally, we confirmed the measured fluidic characteristics of microfluidic channel agreed well with the theoretical calculations. This micro needle can be used both for researches in neuroscience and to treat the brain diseases.

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References