Flexible electronics for bio-signal monitoring in implantable applications

Shusuke Yoshimoto^{a)*}, Teppei Araki, Takafumi Uemura, Yuki Noda, and Tsuyoshi Sekitani*

*Institute of Scientific and Industrial Research, Osaka University,
8-1 Mihogaoka, Ibaraki, Osaka 567-0047, Japan*

a) yoshihy@anken.osaka-u.ac.jp

Abstract: This paper presents flexible electronics with a bio-compatible material interface for bio-signal monitoring applications. Organic thin-film transistors (TFT) are integrated with the bio-compatible gel interface in a 1.2- μm -thick film substrate. An electrocardiogram (ECG) of a rat is amplified by an organic TFT amplifier circuit that is mounted directly on the surface of the heart. In addition to the flexible electronics implantation, an implantable wireless sensor is introduced in this report. Integration between the flexible organic electronics and the silicon devices produces a promising technology for next-generation internet-of-things (IoT) sensors.

Keywords: flexible electronics, organic transistor, implantable device, bio-compatible gel, electrocardiogram, electrocorticography

Classification: Electron devices, circuits and modules

References

- [1] F. Garnier, *et al.*: “All-polymer field-effect transistor realized by printing techniques,” *Science* **265** (1994) 1684 (DOI: [10.1126/science.265.5179.1684](https://doi.org/10.1126/science.265.5179.1684)).
- [2] S. R. Forrest: “The path to ubiquitous and low-cost organic electronic appliances on plastic,” *Nature* **428** (2004) 911 (DOI: [10.1038/nature02498](https://doi.org/10.1038/nature02498)).
- [3] D. H. Kim, *et al.*: “Stretchable and foldable silicon integrated circuits,” *Science* **320** (2008) 507 (DOI: [10.1126/science.1154367](https://doi.org/10.1126/science.1154367)).
- [4] T. Someya, *et al.*: “A large-area, flexible pressure sensor matrix with organic field-effect transistors for artificial skin applications,” *Proc. Natl. Acad. Sci. USA* **101** (2004) 9966 (DOI: [10.1073/pnas.0401918101](https://doi.org/10.1073/pnas.0401918101)).
- [5] Y. Kato, *et al.*: “Large-area flexible ultrasonic imaging system with an organic transistor active matrix,” *IEEE Trans. Electron Devices* **57** (2010) 995 (DOI: [10.1109/TED.2010.2044278](https://doi.org/10.1109/TED.2010.2044278)).
- [6] K. Ishida, *et al.*: “Stretchable EMI measurement sheet with 8×8 coil array, 2 V organic CMOS decoder, and 0.18 μm silicon CMOS LSIs for electric and magnetic field detection,” *IEEE J. Solid-State Circuits* **45** (2010) 249 (DOI: [10.1109/JSSC.2009.2034446](https://doi.org/10.1109/JSSC.2009.2034446)).
- [7] H. Marien, *et al.*: “ADC design in organic thin-film electronics technology on plastic foil,” *Int. Workshop on ADC* (2011).
- [8] W. Xiong, *et al.*: “3-V 6-Bit C-2C digital-to-analog converter using complementary organic thin-film transistors on glass,” *IEEE J. Solid-State*

*Corresponding Authors

- Circuits **45** (2010) 1380 (DOI: [10.1109/JSSC.2010.2048083](https://doi.org/10.1109/JSSC.2010.2048083)).
- [9] H. Marien, *et al.*: “A fully integrated $\Delta\Sigma$ ADC in organic thin-film transistor technology on flexible plastic foil,” IEEE J. Solid-State Circuits **46** (2011) 276 (DOI: [10.1109/JSSC.2010.2073230](https://doi.org/10.1109/JSSC.2010.2073230)).
 - [10] J. Viventi, *et al.*: “Flexible, foldable, actively multiplexed, high-density electrode array for mapping brain activity in vivo,” Nat. Neurosci. **14** (2011) 1599 (DOI: [10.1038/nn.2973](https://doi.org/10.1038/nn.2973)).
 - [11] Q. Qing, *et al.*: “Nanowire transistor arrays for mapping neural circuits in acute brain slices,” Proc. Natl. Acad. Sci. USA **107** (2010) 1882 (DOI: [10.1073/pnas.0914737107](https://doi.org/10.1073/pnas.0914737107)).
 - [12] S. C. B. Mannsfeld, *et al.*: “Highly sensitive flexible pressure sensors with microstructured rubber dielectric layers,” Nat. Mater. **9** (2010) 859 (DOI: [10.1038/nmat2834](https://doi.org/10.1038/nmat2834)).
 - [13] Y. J. Hsu, *et al.*: “A locally amplified strain sensor based on a piezoelectric polymer and organic field-effect transistors,” IEEE Trans. Electron Devices **58** (2011) 910 (DOI: [10.1109/TED.2010.2102631](https://doi.org/10.1109/TED.2010.2102631)).
 - [14] M. D. Angione, *et al.*: “Interfacial electronic effects in functional bilayers integrated into organic field-effect transistors,” Proc. Natl. Acad. Sci. USA **109** (2012) 6429 (DOI: [10.1073/pnas.1200549109](https://doi.org/10.1073/pnas.1200549109)).
 - [15] M. Berggren and A. Richter-Dahlfors: “Organic bioelectronics,” Adv. Mater. **19** (2007) 3201 (DOI: [10.1002/adma.200700419](https://doi.org/10.1002/adma.200700419)).
 - [16] F. Cicoira, *et al.*: “Influence of device geometry on sensor characteristics of planar organic electrochemical transistors,” Adv. Mater. **22** (2010) 1012 (DOI: [10.1002/adma.200902329](https://doi.org/10.1002/adma.200902329)).
 - [17] D. Khodagholy, *et al.*: “Organic electrochemical transistor incorporating an ionogel as a solid state electrolyte for lactate sensing,” J. Mater. Chem. **22** (2012) 4440 (DOI: [10.1039/c2jm15716k](https://doi.org/10.1039/c2jm15716k)).
 - [18] T. Sekitani, *et al.*: “Ultraflexible organic amplifier with biocompatible gel electrodes,” Nat. Commun. **7** (2016) 11425 (DOI: [10.1038/ncomms11425](https://doi.org/10.1038/ncomms11425)).
 - [19] T. C. Huang: “Pseudo-CMOS: A design style for low-cost and robust flexible electronics,” IEEE Trans. Electron Devices **58** (2011) 141 (DOI: [10.1109/TED.2010.2088127](https://doi.org/10.1109/TED.2010.2088127)).
 - [20] T. Yamamoto and K. Takimiya: “Facile synthesis of highly π -extended heteroarenes, dinaphtho[2,3-b:2',3'-f]chalcogenopheno[3,2-b]chalcogenophenes, and their application to field-effect transistors,” J. Am. Chem. Soc. **129** (2007) 2224 (DOI: [10.1021/ja068429z](https://doi.org/10.1021/ja068429z)).
 - [21] K. Fukuda, *et al.*: “Organic pseudo-CMOS circuits for low-voltage large-gain high-speed operation,” IEEE Electron Device Lett. **32** (2011) 1448 (DOI: [10.1109/LED.2011.2161747](https://doi.org/10.1109/LED.2011.2161747)).
 - [22] T. Sekitani, *et al.*: “Flexible organic transistors and circuits with extreme bending stability,” Nat. Mater. **9** (2010) 1015 (DOI: [10.1038/nmat2896](https://doi.org/10.1038/nmat2896)).
 - [23] S. Yoshimoto, *et al.*: “Implantable wireless 64-channel system with flexible ECoG electrode and optogenetics probe,” IEEE Biomedical Circuits and Systems Conference (2016) 476 (DOI: [10.1109/BioCAS.2016.7833835](https://doi.org/10.1109/BioCAS.2016.7833835)).
 - [24] E. C. Leuthardt, *et al.*: “Electrocorticography-based brain computer Interface-the seattle experience,” IEEE Trans. Neural Syst. Rehabil. Eng. **14** (2006) 194 (DOI: [10.1109/TNSRE.2006.875536](https://doi.org/10.1109/TNSRE.2006.875536)).
 - [25] E. C. Leuthardt, *et al.*: “The emerging world of motor neuroprosthetics: A neurosurgical perspective,” Neurosurgery **59** (2006) 1 (DOI: [10.1227/01.NEU.0000221506.06947.AC](https://doi.org/10.1227/01.NEU.0000221506.06947.AC)).
 - [26] E. Castagnola, *et al.*: “PEDOT-CNT-coated low-impedance, ultra-flexible, and brain-conformable micro-ECoG arrays,” IEEE Trans. Neural Syst. Rehabil. Eng. **23** (2015) 342 (DOI: [10.1109/TNSRE.2014.2342880](https://doi.org/10.1109/TNSRE.2014.2342880)).
 - [27] Y. Morikawa, *et al.*: “An origami-inspired ultrastretchable bioprobe film

- device,” IEEE Microelectromechanical Systems (2016) 149 (DOI: [10.1109/MEMSYS.2016.7421580](https://doi.org/10.1109/MEMSYS.2016.7421580)).
- [28] S. Yamagiwa, *et al.*: “Self-curling and -sticking flexible substrate for ECoG electrode array,” IEEE Microelectromechanical Systems (2013) 480 (DOI: [10.1109/MEMSYS.2013.6474283](https://doi.org/10.1109/MEMSYS.2013.6474283)).
- [29] O. Yizhar, *et al.*: “Optogenetics in Neural systems,” Neuron **71** (2011) 9 (DOI: [10.1016/j.neuron.2011.06.004](https://doi.org/10.1016/j.neuron.2011.06.004)).
- [30] B. R. Arenkiel, *et al.*: “In vivo light-induced activation of neural circuitry in transgenic mice expressing channelrhodopsin-2,” Neuron **54** (2007) 205 (DOI: [10.1016/j.neuron.2007.03.005](https://doi.org/10.1016/j.neuron.2007.03.005)).
- [31] B. Schobert, *et al.*: “Halorhodopsin is a light-driven chloride pump,” J. Biol. Chem. **257** (1982) 10306.
- [32] M. Behl, *et al.*: “Shape-memory polymers,” Mater. Today **10** (2007) 20 (DOI: [10.1016/S1369-7021\(07\)70047-0](https://doi.org/10.1016/S1369-7021(07)70047-0)).
- [33] K. Y. Kwon, *et al.*: “Opto-μECoG array: A hybrid neural interface with transparent electrode array and integrated LEDs for optogenetics,” IEEE Trans. Biomed. Circuits Syst. **7** (2013) 593 (DOI: [10.1109/TBCAS.2013.2282318](https://doi.org/10.1109/TBCAS.2013.2282318)).

1 Introduction

The internet of things (IoT) is a broadly attractive topic in many research fields related to clinical, automotive, infrastructure, wearable, and agricultural applications. In the IoT concept, analog signals monitored by a sensor in real space are transferred to cyberspace via the internet. The analog signals can be generated by sensor devices to monitor electrical, magnetical, and physical information.

Flexible electronics are expected to become next-generation sensor devices according to their benefits of low cost, large-area sensing function, and disposability [1, 2, 3]. Physical pressure distribution is monitored by a large-area flexible pressure sensor matrix for artificial skin applications [4]. A flexible ultrasonic imaging sheet can be produced from a polymeric ultrasonic-transducer array and an organic transistor active matrix [5]. Stretchable electromagnetic interference (EMI) measurement sheets have been developed using organic CMOS decoder circuits and 0.18 μm silicon CMOS circuits [6].

The application of flexible electronics has already matured, yielding sensor devices able to generate analog signals. However, current flexible electronics have lower performance than silicon-based technologies can offer. For a flexible electronics society, collaboration with silicon circuits is inevitable in the current performance level to achieve real-time monitoring functions.

Silicon technology presents benefits in terms of high performance, large-scale integration, and low-power operation. Regarding analog-to-digital converters (ADC), flash, SAR, ΔΣ, and ADC have been proposed using organic transistors [7, 8, 9]. However, their resolution and signal bandwidth remain insufficient: less than 7 bit and slower than 1 kHz. Because of the constrained capabilities of the flexible electronics, it is important to combine the benefits of flexible electronics and silicon technology, as presented in Fig. 1.

Sensing using a flexible interface imposes little stress to objects because of its mechanical flexibility. The generated signal can be amplified by the flexible circuit

that is directly integrated in the sensing terminal. The amplified signals are then digitized and transferred to the internet using silicon technology.

Fig. 1. Integration of flexible electronics and silicon technology for internet of things (IoT). The sensing interface and amplifying functions are achieved in the flexible electronics. High-performance functions such as ADC and data transmission are based on the silicon technology.

As described in this report, we introduce the combination of flexible electronics and silicon technology for the application of implantable (*in vivo*) devices. *In vivo* electronic monitoring systems represent a promising technology for obtaining biosignals with high spatiotemporal resolution. Flexible electronic devices have attracted much attention because they are applicable on complex curved surfaces with a large area of coverage. Recently, flexible *in vivo* neural interfaces have been demonstrated using silicon nanomembrane thin-film transistor (TFT) active matrices to monitor neural activities in a cat brain electrically with high spatial and temporal resolution and to elucidate and map neural circuits in acute brain slices [10, 11]. Organic TFTs and related structures have been investigated to improve mechanical flexibility further [12, 13, 14, 15, 16, 17].

2 Flexible organic amplifier with biocompatible gel composite

This section explains the fabrication of a biocompatible highly conductive gel composite to amplify weak bio-signals: it comprises a multi-walled carbon nano-tube (CNT) dispersed sheet and a mechanically flexible organic active matrix amplifier on a 1.2- μm -thick polyethylene-naphthalate film [18]. A 3-D integrated image is presented in Fig. 2, showing the bio-compatible conductive gel and CNT-sheet composite, thin-film capacitor, organic pseudo-CMOS inverter amplifier array [19], and organic TFT active matrix. The bio-compatible gel interface can be attached directly to an organ of an animal without antibody response. The integrated flexible organic amplifier circuit increases the bio-signal amplitude near the electrode surface.

Fig. 2. Vertically integrated bio-signal amplifier circuit matrix sheet. CNT-sheet and bio-compatible gel composite, thin-film input capacitor, organic pseudo-CMOS amplifier array, and organic TFT active matrix are developed in the flexible sheet. Adapted from [18]. Copyright 2016, Nature Publishing Group.

Fig. 3(a) portrays high-resolution cross-sectional electron microscopy images of a CNT/polyrotaxane gel composite. The transmission electron microscopy (TEM) images of the specimen after drying in vacuum clearly depict the polyrotaxane-based gels as coated onto the multiwalled CNTs with 5–10 nm diameter. A bilayer CNT-sheet/gel composite was formed by 365-nm UV irradiation for 5 min, as presented in Fig. 3(b).

The entire CNT surface is uniformly coated with aqueous polyrotaxane-based gel using a hydrophilic ionic liquid. This structure, which has an increased effective surface ratio (electrode area), exhibits admittance of 100 mS/cm^2 without sacrificing softness, which is an important characteristic of gels. The chemically stable gel composite is mechanically flexible and stretchable, with Young's modulus of approximately of approximately 10 kPa. When the gel composite was implanted into the hypodermal tissue of a living body for four weeks, the foreign-body reaction of the tissue was less than that obtained when using existing metal electrodes.

The bilayer gel composite had admittance of 100 mS/cm^2 , which is the highest reported value among CNT-based conductive gels, even in a low-frequency range. The admittance of the CNT/polyrotaxane gel composite was evaluated using AC measurements, as shown in Fig. 3(c). Polyrotaxane-based gels with different conductive layers, such as a graphite sheet, Au-coated film, and Al-coated film, are also shown for comparison. The CNT-sheet/gel composite showed the highest admittance in the low-frequency range because of the large surface ratio of the CNT-based conductive electrodes, which have admittance of two or three orders of magnitude higher than those of metal/gel-based composite electrodes. In fact, the CNT/gel interface formed a highly capacitive electronic double layer with capacitance exceeding $140 \mu\text{F/cm}^2$. We organized all biocompatibility tests under rules of internationally standardized procedures for colony-forming assay (ISO10993-5) and implant assay (ISO10993-6).

Fig. 3. (a) TEM image of CNT/gel composite. (b) The bilayer CNT-sheet/gel formed by 365-nm UV irradiation for 5 min. (c) Measured admittance of the CNT/gel composite, graphite sheet, Au-coated film, and Al-coated film. Adapted from [18]. Copyright 2016, Nature Publishing Group.

For this study, a flexible organic amplifier is integrated into the CNT-sheet/gel interface. The amplifier is formed by an inverter with a pseudo-CMOS layout comprising four p-channel organic transistors with semiconducting dinaphtho-[2,3-b:2',3'-f]thieno[3,2-b]thiophene (DNNT) [20] and an organic self-assembled monolayer (SAM) gate dielectric [21], as portrayed in Fig. 4(a). A waterproof hybrid encapsulation stack comprising a 200-nm-thick Au layer sandwiched between a 100-nm-thick or a 1.2- μm -thick parylene layer was deposited on the transistors to serve as a passivation layer against oxygen diffusion, humidity, and mechanical attrition *in vivo* and as an overcoat layer to enhance the mechanical flexibility and durability because the organic semiconducting layer was located at the neutral strain position [22], leading to reduction in the effective strain at the TFT plane. The ultrathin transistors, which exhibited high mechanical flexibility, showed no change in performance even with a 50- μm bending radius, as portrayed in Fig. 4(b). Fig. 4(c) shows the two-dimensional array of organic amplifiers fabricated directly on a 1.2- μm -thick flexible polyethylene-naphthalate (PEN) substrate.

Fig. 4. (a) An organic p-channel organic transistor with semiconducting DNNT and SAM gate dielectric, (b) current characteristics before, during, and after bending the flexible transistor for 50- μm bending radius, and (c) two-dimensional array of organic amplifier fabricated on a 1.2- μm -thick PEN film substrate. Adapted from [18]. Copyright 2016, Nature Publishing Group.

Fig. 5(a) presents a circuit diagram and a photograph of the organic amplifier comprising a pseudo-CMOS inverter on a 1.2- μm -thick ultraflexible PEN with organic circuits, CNT-gel composite (CNT-gel), input capacitor (C), and resistance (R). The input impedance and power requirements for the amplifiers are, respectively, 100 k Ω and 10 μW . The pseudo-CMOS inverter has five electrical terminals: input, output, power source voltage V_{DD} , tuning voltage V_{SS} , and ground GND. It can operate within 2 V with the signal gain exceeding 400. Moreover, it can maintain this level of electrical performance even when bent at a bending radius of 50 μm . The propagation delay per stage of the transistors was 23.4 μs at 2 V, which corresponds to the frequency response of 42.7 kHz on a single transistor.

A direct epicardial electrocardiogram was obtained using self-feedback-type organic amplifiers by combining the inverter, a conductive gel sheet, a resistor made of conductive carbon paste ($R = 2.1 \text{ M}\Omega$), and a capacitor formed using SAM/ AlO_x dielectrics sandwiched between Au electrodes ($C = 670 \text{ nF}$). The gel composite was placed on a rat heart and was interconnected to an organic amplifier sheet using electrical connections as presented in Fig. 5(b). A 1.2-mV input signal obtained from the heart was amplified to a 220-mV output signal using an amplifier (signal gain: approx. 200) as presented in Fig. 5(c). The input SNR of 0.53 was increased to 64 after amplification.

This report is the first to describe a demonstration of *in vivo* bio-signal amplification from hypodermal tissue using organic circuits. Moreover, this report is the first to describe a technique for integration of an ultrathin active matrix amplifier array with biocompatible gels. This report also includes the first demonstration of electrocardiogram measurement of the heart of a living rat.

Fig. 5. (a) Image and schematic of the fabricated pseudo-CMOS amplifier circuit with a 2.1 M Ω resistor and a 670 nF capacitor. (b) The flexible amplifier is directly attached to a rat's heart. (c) Original and amplified rat ECG signals. Adapted from [18]. Copyright 2016, Nature Publishing Group.

3 Implantable wireless 64-channel system with flexible ECoG electrode and optogenetics probe

This section introduces an implantable wireless sensor system including 64-channel 24-bit ADC, 2-channel LED drivers for optogenetics [23], and WiFi wireless communication with collaboration of a bio-compatible and flexible electrocorticography (ECoG) electrode.

Recently, ECoG has become an attractive tool for clinical observations, medical research, and Brain–Machine Interface (BMI) studies [24]. Its *in vivo* approach presents benefits for spatial resolution, signal-to-noise ratio, and frequency range compared to scalp electroencephalography (EEG) [25]. Flexible sheet electrodes reportedly achieve less-invasive ECoG recording with soft materials [26, 27, 28]. Flexible sheet electrodes also provide biocompatibility and mechanical softness, which can facilitate and enhance long-term monitoring. Along with ECoG recording, a manipulation method of neurons, optogenetics, has become widely recognized as a tool to activate and inhibit cells into which channelrhodopsin has been introduced [29]. Channelrhodopsin-2 (ChR2) can be used for millisecond control of specific neurons by blue light [30]. Halorhodopsin is a light-gated pump by yellow light [31]. The neural manipulation function can be implemented in small systems using recently developed low-power devices.

The system proposed in this report is designed to realize long-term ECoG monitoring and remote optogenetics for primate animals during free movements. The proposed sensor implantation image is depicted in Fig. 6. The flexible electrode is covered by a bio-compatible gel interface that is attached directly to the primate's brain. The sensor system must be made smaller, thinner, and lighter because primates cannot behave normally with heavy or large monitoring devices attached to them. The proposed system has been minimized with the ADC stacked structure with low-noise (1- μ V input referred noise) and 24-bit high resolution. In addition to the implantable system, this section presents a flexible ECoG recording electrode, flexible LED probe, and a software interface for the system.

Fig. 6. Flexible ECoG recording electrode and flexible LED probe for optogenetics are implanted into a primate's skull. The electrode and probe are connected to a wireless system implanted in the primate's back. Adapted from [23]. Copyright 2016, IEEE.

The 75 mm $\times$ 40 mm $\times$ 12 mm main board, weighing 33.53 g, was developed to include a Li-ion battery. It and a 25 mm $\times$ 20 mm $\times$ 2.5 mm, 8-channel, 24-bit ADC board weighing 0.98 g are presented respectively in Figs. 7(a) and 7(b). The main board has eight connectors for the ADC boards to accommodate 64-channel analog inputs. The main board, with its Wi-Fi communication module and FPGA controller, can realize 64-channel 24-bit 1-kHz sampling ($64 \times 24 \times 1000 = 1.536$ Mbit/s) without data compression. The 1400-mAh li-ion battery can be charged

using a micro-USB port in the main board. The board also has two LED drivers that can output up to 10 mA driving current. The ADC board has an 8-channel 24-bit analog front end and a reference voltage (= supply voltage/2) generator using an operational amplifier. The main and ADC boards are connected by 23-pin 0.3-mm-pitch flexible cables (0.232 g/cable). Eight ADC boards are stacked to realize an ADC cube architecture that in a small device: 25 mm × 20 mm × 20 mm and 7.82 g. Fig. 7(c) presents a block diagram of the proposed system. The main board DC–DC converter provides 5-V and 3-V VDD. The Wi-Fi module and FPGA operate at 5-V VDD; the ADC needs both VDDs.

Fig. 7. (a) The main board consists of WiFi communication, an LED driver, a USB charger, an FPGA controller, and eight connectors for ADC boards. (b) Each ADC board has eight analog inputs. Eight ADC boards are stacked to achieve the 24-b 64-ch ADC function. (c) Overall block diagram for the main board and ADC board. Adapted from [23]. Copyright 2016, IEEE.

Fig. 8 presents the developed flexible sheet electrode with a parylene substrate and the flexible 50- μ m-thick LED probe and a 500- μ m-wide polyimide substrate. The flexible sheet electrode is covered by a shape-memory polymer designed to be flexible from solid state at 35°C body temperature [32]. The electrode is solid at room temperature when inserted into the brain, but it is flexible because of the body temperature for non-invasive measurement. The LED cable includes 465-nm 3.3-V blue LED and 590-nm 2.5-V yellow LED. The LEDs have 1.0 mm × 0.5 mm × 0.5 mm size and 120° directivity. The LEDs have 1.58-mW/mm² irradiance at 2.89-V and 5-mA power consumption, which can induce action potentials [33].

Fig. 8. Flexible ECoG electrode and LED probe for optogenetics. The stimulation probe and recording electrode are implanted directly in a skull of an animal for communication with the brain. Adapted from [23]. Copyright 2016, IEEE.

All methods of animal care and use were performed by medical doctors with approval from the Institutional Animal Care and Use Committee of Osaka University. A rat (350 g, 20 week old male) was anesthetized and maintained with isoflurane in a 50% mixture of oxygen and air. For the experiments, the temperature of the rat was maintained at 36°C. Fig. 9 presents an ECoG recording with a flexible sheet electrode, the main board, and an ADC board. The electrode sheet has a ground electrode, a reference electrode, and sensing electrodes. The ground and reference electrodes were placed at different places on the brain surface. The sensing electrodes are inserted in the brain to monitor ECoG signals. The proposed system recorded the ECoG signals.

Fig. 9. (a) Experimental setup for the implanted system to confirm the stable recording function of this system. (b) The ECoG signal is recorded by a rat's brain using the flexible electrode and the implantable system. Adapted from [23]. Copyright 2016, IEEE.

4 Conclusion

As described herein, using the novel gelatinous composite as a biocompatible electrode for implantable electronics, we have reduced the foreign-body reaction compared with widely used metal electrodes, whereas the new material maintains extraordinarily high AC admittance of 100 mS/cm², even in the low-frequency range. The gel composites described above were integrated with a two-dimensional

array of ultrathin flexible organic amplifiers to produce biocompatible interfaces between the biological tissues and the electrodes of the electronic components. The feasibility of the system was demonstrated through direct measurement of epicardial electrocardiogram signals at an amplification factor of 200.

This report also describes a wireless 64-channel electrocorticogram (ECoG) recording system with two-channel optogenetics manipulation. The proposed system, which is designed to be extremely small to make it implantable into an animal, comprises a main board and ADC boards. A biocompatible and flexible ECoG electrode is proposed for non-invasive monitoring. The electrode is covered by shape-memory polymer, which is solid at room temperature (27°C) and which is flexible at body temperature (35°C). In addition to the recording channels, two blue and yellow LEDs were developed on a flexible polyimide cable for optogenetics. This report presents experimentally obtained results demonstrating that the proposed system can record a rat ECoG with a flexible electrode and the proposed wireless system.

Acknowledgments

This work was partially conducted by Prof. T. Someya and his coworkers at The University of Tokyo, Prof. M. Hirata and his coworkers at Osaka University, and Dr. T. Suzuki of the National Institute of Information and Communications Technology (NICT). This work was supported by the Brain Mapping by Integrated Neurotechnologies for Disease Studies (Brain/MINDS) project from the Japan Agency for Medical Research and Development (AMED), a grant for “Development and application of an implantable device for brain-machine interfaces using high-speed wireless data transfer and big data analyses of neural information”, the Commissioned Research of NICT, Japan, the Center of Innovation Program from Japan Science and Technology Agency (JST), and KAKENHI (16K21141) grant funded by the Japan Society for the Promotion of Science (JSPS).

Shusuke Yoshimoto

received B.E. and M.E. degrees in Computer and Systems Engineering from Kobe University, Hyogo, Japan, in 2009 and 2011, respectively. He earned a Ph.D. degree in Engineering from the university in 2013. He was a JSPS Research Fellow during 2013–2014. He worked at the Department of Electrical Engineering of Stanford University as a post-doctoral researcher during 2013–2015. Since 2015, he has been an Assistant Professor at The Institute of Scientific and Industrial Research at Osaka University.

Teppei Araki

received M.E. and Ph.D. degrees from Department of Adaptive Machine Systems of Osaka University, Osaka, Japan in 2011 and 2014, respectively. He was a JSPS Research Fellow during 2012–2014. He studied at Holst Centre in the Netherlands for a year during the doctoral program. Since 2014, he has been an Assistant Professor at The Institute of Scientific and Industrial Research at Osaka University.

Takafumi Uemura

received B.S., M.S., and Ph.D. degrees in applied physics from Osaka University, Osaka, Japan, in 2003, 2005, and 2008, respectively. During 2008–2010, he was a Postdoctoral Researcher at Osaka University. During 2010–2013, he was an Assistant Professor of the Institute of Scientific and Industrial Research, Osaka University. During 2013–2015, he was a Project Lecturer at The University of Tokyo. During 2013–2014, he worked at IMEC as a Visiting Professor at KU Leuven, Belgium. Since 2015, he has been a Specially Appointed Associate Professor at Osaka University.

Yuki Noda

received a B.S. degree from Nihon University in 2006. He earned M.S. and Ph.D. degrees in Division of Environmental Science, Graduate School of Environmental Science, Hokkaido University in 2008 and 2011, respectively. He was a JSPS Research Fellow during 2009–2011. During 2011–2016, he was a postdoctoral researcher at the National Institute of Advanced Industrial Science and Technology. Since 2016, he has been a specially appointed Assistant Professor at The Institute of Scientific and Industrial Research at Osaka University.

Tsuyoshi Sekitani

received a B.S. degree from Osaka University, Osaka, Japan in 1999, and a Ph.D. degree in applied physics from The University of Tokyo, Tokyo, Japan in 2003. During 2003–2010, he was a Research Associate. In 2011, he was an Associate Professor at the School of Engineering of The University of Tokyo. In 2014, he became a Professor in The Institute of Scientific and Industrial Research at Osaka University. In 2017, he was promoted to Distinguished Professor at Osaka University.