Wireless Sensor Readout System for Bone Intramedullary Pressure Monitoring Applications

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2021 IEEE Sensors | 978-1-7281-9501-8/21/$31.00 ©2021 IEEE | DOI: 10.1109/SENSORS47087.2021.9639485

Abstract—**Wireless pressure sensors have attracted significant research interests in recent years, especially in the field of biomedical applications. Although majority of wireless pressure sensors utilize inductive coupling technology, these devices are limited to operation range of millimeters to centimeters. In this work, we report a wirelessly operated pressure sensing system based on Bluetooth® Low Energy (BLE) for long-range implantable intramedullary pressure monitoring. Pressure measurements of the fabricated pressure sensing system showed that the pressure sensor provides timely and accurate responses to applied pressure ranging from 0-150 mmHg. Wireless transmission range measurement showed that the received signal power level ranges from -60 dBm to -75 dBm in a distance range of 0.5 m to 8 m. With the maximum sensitivity of -97 dBm, the wireless pressure sensing range is expected to be capable of long-range transmission in typical indoor operation.**

Keywords—wireless; pressure sensor; in vivo; intramedullary

I. INTRODUCTION

Implantable devices have played a significant role in advancing MEMS sensing technologies into emerging biomedical fields in recent years. Innovative implantable sensors have been demonstrated to monitor numerous physiological signals including pressure [1], [2], strain [3], flow [4], *p*H [5], oxygen [6], glucose [7], etc. Pressure sensors, among others, have attracted the most effort as pressure is one of the most vital physiological signals. Various studies have demonstrated implantable pressure sensors for monitoring intravascular [1], intraocular [8], and intracranial [9] pressure, which may be indicatives of cardiovascular, ocular, or cerebral diseases, respectively.

Despite the tremendous progress, the development of implantable pressure sensors is still faced with a few technical challenges in terms of biocompatibility, comfortability, and invasiveness. These challenges have thus triggered the exploration of new materials, design, and techniques to develop implantable pressure sensors. Novel materials, including polymers (i.e., polyimide, parylene, SU-8, etc.), carbon nanotube (CNT), gold/silver nanowires (NWs), and gallium based liquid metals have been utilized to fabricate flexible pressure sensors [10]–[12] with improved biocompatibility and comfortability.

Nonetheless, minimizing the invasiveness of these implantable pressure sensors remains one of the major challenges. Significant efforts have been directed to develop wireless implantable pressure sensors. Vast majority of current wireless pressure sensors utilize inductive coupling for wireless power transfer and/or sensing including our previous works [8], [13], [14]. Despite the compactness and reduced invasiveness of these passive devices, they suffer from major limitations as well. The efficiency of inductive coupling depends on the distance and collinear alignment between the external coil and the receiver coil located inside the implantable devices. The operating distance is limited to a range of only millimeters to at most centimeters under collinear positioning. Due to the movement of muscles or tissues and other anatomical restrictions, it could be daunting to obtain readout externally without knowing the exact position of the implanted device. Furthermore, these devices can only operate with an external antenna to transfer power and/or data wirelessly. This has hindered their applications for continuous health monitoring with mobile phones or general computing devices. Therefore, a wireless implantable pressure sensor capable of long-range transmission and interfacing with widely available wireless communication protocols (i.e., Wi-Fi or Bluetooth®) is desirable for certain health monitoring or *in vivo* study purposes.

In our previous work, a wirelessly operated implantable fluid modulator was developed for *in vivo* studies of intramedullary fluid modulation and its efficacy in bone density enhancement for rat model [15]. However, a wireless pressure sensor is still required to elucidate the correlation of induced intramedullary pressure and bone growth. So far, most existing *in vivo* studies utilized commercial pressure sensors that require catheters surgically inserted through skins [16], [17]. Such invasive methodologies impose significant risks of infections and restrictions on the normal physical activities of the test subjects. Aiming to reduce the invasiveness of real-time intramedullary pressure monitoring, a wireless implantable pressure sensor with long range transmission capability enabled by Bluetooth® Low Energy (BLE) is presented to complement the lack of viable options of tools for *in vivo* studies of intramedullary pressure and fluid modulation.

II. SYSTEM DESIGN

Fig. 1 shows the working principle of the wireless pressure sensing system with insets showing the schematic diagrams of the wireless pressure sensor and the external wireless transceiver module. The wireless pressure sensor is designed to monitor real-time pressure in the femora of Fischer-344 rats induced by on-demand intramedullary pressure modulation. Because of the available sub-dermal volume, the wireless pressure sensor would be placed in the back of the rats. The intramedullary cavity of the femur is connected to the pressure sensor unit through polyethylene (PE) tubing filled with 0.9% heparinized saline to prevent blood clotting.

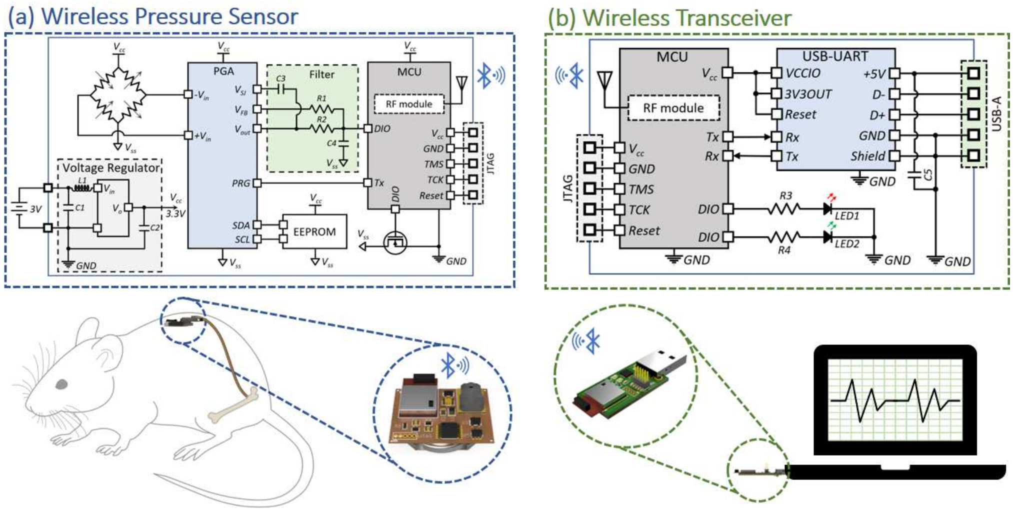

Fig. 1. Schematics and working principle of the wireless pressure sensing system: (a) Wireless pressure sensor, (b) Wireless transceiver module.

As shown in Fig. 1(a), the wireless sensor consists of three functional components: a voltage regulator, a pressure sensing unit as well as amplifying circuit, and a microcontroller integrated with RF module for wireless communications. The pressure sensing unit converts applied physiological pressure to voltage differences by a temperature-compensated Wheatstone bridge. The voltage signal is then amplified and filtered by the gain amplifier circuit before converting to digital readings that can be transmitted wirelessly to external transceiver. Fig. 1(b) shows the architecture of the wireless transceiver module. The transceiver acts as a wireless gateway to interface with the wireless pressure sensor and a PC terminal. Two-way data communication is processed by a USB (universal serial bus)-to-UART (universal asynchronous receiver-transmitter) serial interface IC (FT231XQ-R, FTDI, Glasgow, UK).

Fig. 2 shows the architecture of the software for the wireless pressure sensing system. The software for the whole wireless pressure sensing system includes three parts: Graphic User Interface (GUI) in the PC terminal, and firmware for the wireless transceiver module and pressure sensor. The GUI functions as the main controller to send commands and receive data for visualization and recording purposes. The firmware of the wireless transceiver is configured to convert data in different formats to bridge serial communications and wireless communications based on Bluetooth®, while that of the wireless pressure sensor is responsible for processing the received commands and sending the data wirelessly.

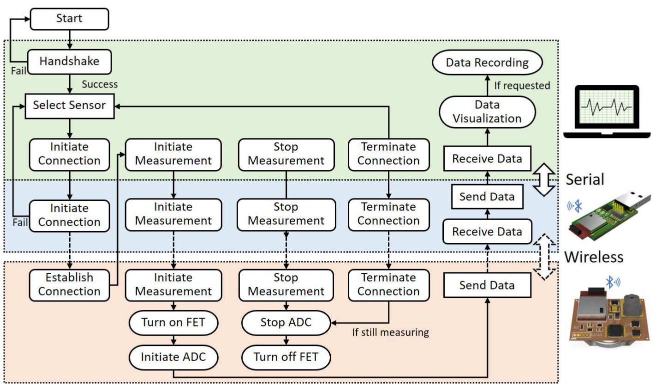

Fig. 2. Software architecture of the wireless pressure sensing system.

III. Fabrication and Assembly

The circuit boards for the hardware of the wireless pressure sensing system were designed and fabricated. The board for the wireless pressure sensor was made by using polyimide (PI) for its flexibility, while that of the transceiver with standard FR4 substrate. The electronic components were soldered manually by using solder paste (MG Chemicals, Ontario, Canada) and a rework station with hot gun set to be 300 °C. Due to the small size of surface-mount devices (SMD), a solder mask was used to assist the application of solder paste to avoid solder bridging or insufficient wetting. The pressure sensing unit was soldered separately using back side heating to prevent the low-melting point (~110 °C) polycarbonate (PC) housing from melting. Fig. 3 shows the optical images of the assembled wireless pressure sensor and transceiver module. The size of the assembled wireless pressure sensor was measured to be approximately 28 × 26 × 8 mm. After the assembly, the inlet of the pressure sensing unit was bonded to PE90 tubing (ID: 0.86 mm, OD: 1.27 mm) with a 3D-printed biocompatible polylactic acid (PLA) [18] adapter by using medical grade epoxy (M-31CL, Henkel Loctite, Düsseldorf, Germany).

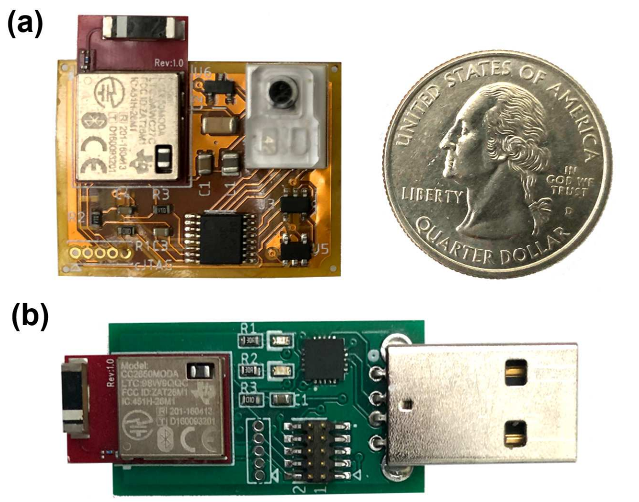

Fig. 3. Optical images of (a) assembled wireless pressure sensor, and (b) assembled wireless transceiver module in comparison to a US quarter coin.

IV. CHARACTERIZATIONS

A. Pressure Response

Since the relevant intramedullary pressure under on-demand modulation typically ranges from 0 to 80 mmHg, the wireless pressure sensor was calibrated in the range of 0 to approximately 150 mmHg for safe operation. The calibration setup utilized an additional wireless MCU board (CC2650 LaunchPad, Texas Instruments, Dallas, TX, USA), commercial pressure sensor, 3D-printed adapter and connecting tubing. Fig. 4(a) shows the schematic diagram of the calibration setup.

The calibration coefficients for the amplifier were calculated to achieve linear and accurate voltage output with respect to pressure. With amplification factor of approximately 467.5, the analog output sensitivity of the wireless pressure is 7.714 mV/mmHg, while the built-in analog-to-digital converter of the MCU is capable of 1.050 mV resolution. Fig. 4 summarizes the typical pressure calibration results in comparison to the reference pressure sensor. The temporal pressure response of the wireless pressure sensor agrees well with the referenced pressure in the relevant pressure range of -20 mmHg to 150 mmHg. The pressure output error was calculated by subtracting the pressure reading of the wireless sensor and the reference sensor. As shown in the inset of Fig. 4(b), output error ranges from approximately −1 mmHg in the low-pressure region to ~3 mmHg in the high-pressure region with a few outliers. These outliers are mostly contributed by superposition of noises from both sensors. The actual error of the wireless pressure sensor alone is expected to be lower. Fig. 4(c) plots the pressure reading from the wireless pressure sensor against that of the reference sensor. The pressure response of the wireless pressure sensor is linear with respect to applied pressure. The wireless pressure sensor exhibits a minor deviation from ideal response $P_{DUI} = P_{ref}$ by approximately 2.34%.

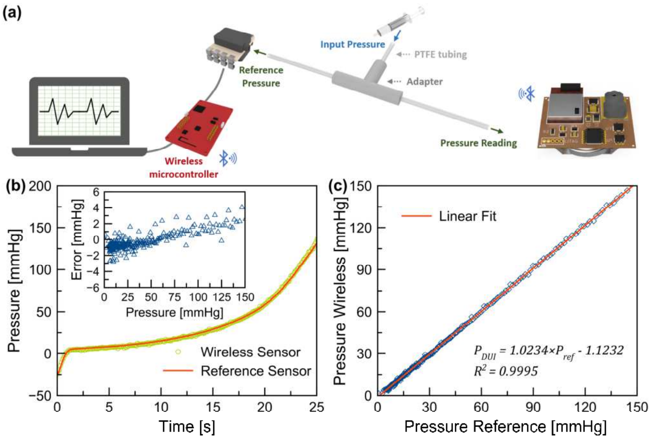

Fig. 4. (a) Calibration setup with a commercial pressure sensor as reference, (b) temporal pressure response of the wireless sensor and reference sensor with the inset showing the error with respect to applied pressure, and (c) pressure response of the wireless sensor (Y axis) plotted against that of the reference sensor (X axis) with linear fit.

B. Range Measurement

The Received Signal Strength Indication (RSSI) is commonly used to represent the power level of signal between two wireless RF devices. RSSI is dependent on the distance between the devices, as electromagnetic wave propagates following the inverse square law. The relationship between RSSI and distance is given as [19]

$$P_{RSSI} = -10\alpha\log_{10}(d) - P_0 \quad (1)$$

where P_{RSSI} is the power level of received signal in dBm, P_0 is the received power level at 1 m reference distance in dBm, α is the path loss coefficient, and d is the distance between the wireless devices.

The RSSI measurement was carried out by using the CC2650LaunchPad at transmission power of 0 dBm (1 mW). Measurements were performed indoor from 0.5 m to 8 m without wall obstructions. Fig. 5 summarizes the RSSI measurement results and fitted according to (1). The results show that the received signal of the wireless pressure sensor ranges from -61.2 dBm at 0.5 m to -75 dBm at 8 m, which is still within the maximum sensitivity of the wireless MCU of -97 dBm [20]. The RSSI measurement result shows that the wireless pressure sensor can achieve an operating range of 8 m placed in indoor environment.

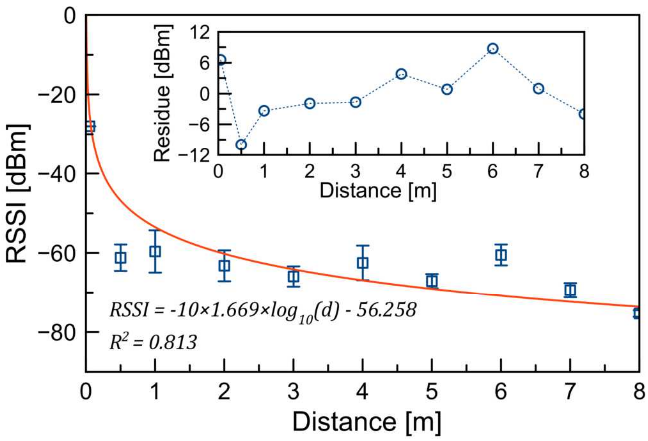

Fig. 5. Received Signal Strength Indication (RSSI) of the wireless pressure sensor, with the inset showing the residue from curve fitting.

V. CONCLUSIONS

We report the design and fabrication of an active wireless pressure sensing system based on Bluetooth® Low Energy (BLE) for *in vivo* bone intramedullary pressure monitoring. Pressure response measurement results show that the wireless pressure sensor was calibrated with a minor deviation from referenced pressure by approximately 2.4%. Wireless transmission range measurement shows that the received signal power level measured indoors ranges from -60 dBm to -75 dBm in a distance range of 0.5 m to 8 m. With the maximum sensitivity of -97 dBm, the wireless pressure sensing range is expected to be 8 m in indoor environment. These results confirm that the wireless pressure sensing system is fully functional and can be used to monitor pressure wirelessly in *in vivo* studies of bone intramedullary pressure modulation or other physiological pressure sensing using implantable biomedical devices.

ACKNOWLEDGMENT

The authors would like to acknowledge the U.S. National Science Foundation (NSF) for financial support for this study under the grant NSF ECCS-1710824.

REFERENCES

[1] C. C. Yeh, S. H. Lo, M. X. Xu, and Y. J. Yang, "Fabrication of a flexible wireless pressure sensor for intravascular blood pressure monitoring," *Microelectron. Eng.*, vol. 213, pp. 55–61, 2019.

[2] S. A. Bello and C. L. Passaglia, "A Wireless Pressure Sensor for Continuous Monitoring of Intraocular Pressure in Conscious Animals," *Ann. Biomed. Eng.*, vol. 45, no. 11, pp. 2592–2604, 2017.

[3] F. Umbrecht, M. Wendlandt, D. Juncker, C. Hierold, and J. Neuenschwander, "A wireless implantable passive strain sensor system," *Proc. IEEE Sensors*, vol. 2005, pp. 20–23, 2005.

[4] J. H. Cheong *et al.*, "An inductively powered implantable blood flow sensor microsystem for vascular grafts," *IEEE Trans. Biomed. Eng.*, vol. 59, no. 9, pp. 2466–2475, 2012.

[5] H. Cao *et al.*, "An implantable, batteryless, and wireless capsule with integrated impedance and pH sensors for gastroesophageal reflux monitoring," *IEEE Trans. Biomed. Eng.*, vol. 59, no. 12–2, pp. 3131–3139, 2012.

[6] L. Rivas *et al.*, "Micro-needle implantable electrochemical oxygen sensor: ex-vivo and in-vivo studies," *Biosens. Bioelectron.*, vol. 153, 2020.

[7] S. D. Psoma, P. D. van der Wal, O. Frey, N. F. de Rooij, and A. P. F. Turner, "A novel enzyme entrapment in SU-8 microfabricated films for glucose micro-biosensors," *Biosens. Bioelectron.*, vol. 26, no. 4, pp. 1582–1587, 2010.

[8] N. Xue, S. P. Chang, and J. B. Lee, "A SU-8-based microfabricated implantable inductively coupled passive RF wireless intraocular pressure sensor," *J. Microelectromech. Syst.*, vol. 21, no. 6, pp. 1338–1346, 2012.

[9] M. W. A. Khan, T. Bjorninen, M. Rizwan, L. Sydanheimo, and L. Ukkonen, "Piezoresistive pressure sensor for ICP monitoring: Remote powering through wearable textile antenna and sensor readout experiment," in *2016 IEEE Antennas and Propagation Society International Symposium, APSURSI 2016 - Proceedings*, 2016, pp. 997–998.

[10] H. Zhang *et al.*, "Piezoresistive Sensor with High Elasticity Based on 3D Hybrid Network of Sponge@CNTs@Ag NPs," *ACS Appl. Mater. Interfaces*, vol. 8, no. 34, pp. 22374–22381, 2016.

[11] K. Wang *et al.*, "Soft gold nanowire sponge antenna for battery-free wireless pressure sensors," *Nanoscale*, vol. 13, no. 7, pp. 3957–3966, 2021.

[12] M. G. Kim, H. Alrowais, and O. Brand, "3D-Integrated and Multifunctional All-Soft Physical Microsystems Based on Liquid Metal for Electronic Skin Applications," *Adv. Electron. Mater.*, vol. 4, no. 2, p. 1700434, 2018.

[13] S. H. Cho, N. Xue, L. Cauller, W. Rosellini, and J. B. Lee, "A SU-8-based fully integrated biocompatible inductively powered wireless neurostimulator," *J. Microelectromech. Syst.*, vol. 22, no. 1, pp. 170–176, 2013.

[14] N. J. Cleven *et al.*, "A novel fully implantable wireless sensor system for monitoring hypertension patients," *IEEE Trans. Biomed. Eng.*, vol. 59, no. 12 PART2, pp. 3124–3130, 2012.

[15] Z. Chen, S. Noh, R. D. Prisby, and J. B. Lee, "Implanted Wireless Intramedullary Fluid Modulator for Bone Density Augmentation," in *Proceedings of the IEEE International Conference on Micro Electro Mechanical Systems (MEMS)*, 2020, January, pp. 384–387.

[16] Y. X. Qin and H. Lam, "Intramedullary pressure and matrix strain induced by oscillatory skeletal muscle stimulation and its potential in adaptation," *J. Biomech.*, vol. 42, no. 2, pp. 140–145, 2009.

[17] M. Hu, J. Cheng, and Y. X. Qin, "Dynamic hydraulic flow stimulation on mitigation of trabecular bone loss in a rat functional disuse model," *Bone*, vol. 51, no. 4, pp. 819–825, 2012.

[18] J. M. Anderson and M. S. Shive, "Biodegradation and biocompatibility of PLA and PLGA microspheres," *Adv. Drug Deliv. Rev.*, vol. 64, no. SUPPL., pp. 72–82, 2012.

[19] N. Patwari, J. N. Ash, S. Kyperountas, A. O. Hero, R. L. Moses, and N. S. Correal, "Locating the nodes: Cooperative localization in wireless sensor networks," *IEEE Signal Process. Mag.*, vol. 22, no. 4, pp. 54–69, 2005.

[20] M. Magno, F. Vultier, B. Szebedy, H. Yamahachi, R. H. R. Hahnloser, and L. Benini, "A Bluetooth-Low-Energy Sensor Node for Acoustic Monitoring of Small Birds," *IEEE Sens. J.*, vol. 20, no. 1, pp. 425–433, 2020.