A Wireless Self-Powered Urinary Incontinence Sensor System

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Abstract: A self-powered urinary incontinence sensor system consisting of a urine-activated coin battery and a wireless transmitter has been developed as an application for wireless biosensor networks. The urine-activated battery makes possible both the sensing of urine leakage and self-powered operation. An intermittent power-supply circuit that uses an electric double-layer capacitor (EDLC) with a small internal resistance suppresses the supply voltage drop due to the large internal resistance of the battery. This circuit and a 1-V surface acoustic wave (SAW) oscillator reduce the power dissipation of a wireless transmitter. The SAW oscillator quickly responds to the on-off control of the power supply, which is suitable for intermittent operation. To verify the effectiveness of the circuit scheme, the authors fabricated a prototype sensor system. When the volume of urine is 0.2 ml, the battery outputs a voltage of over 1.3 V; and the sensor system can transmit signals over a distance of 5 m.

Key Words: wireless, self-powered, biosensor, urine-activated battery, ultralow-power.

1. Introduction

Healthcare services are experiencing increasing demand for personal health monitoring systems and using biosensors. One type of biosensor monitors the composition of a bodily fluid, such as blood or urine. Two key requirements that make a biosensor easy to use are a wireless capability, which enables a person to move about freely, and self-powered operation, which eliminates the need for periodic battery replacement. The authors have already developed a self-powered wireless system [1] that consists of a 0.5-V digital LSI, a 1-V analog/RF LSI, and a DC-DC converter and has a power source consisting of solar cells and a thermoelectric generator. The research showed that self-powered operation requires the development of two key elements. One is an ultralow-power module with a power dissipation at the 1-mW level, which is achieved by lowering the supply voltage down to around 1 V. The other is a micropower generator that produces a power of over 1 mW from an ambient energy source, such as the light, movement, and thermal energy around us.

A urine-activated battery [2] is one type of micropower generators. It outputs a voltage of 2.5 V without an output load. However, when the battery is directly connected to a wireless transmitter with an equivalent impedance of 1 kΩ, its large internal resistance causes the supply voltage to the transmitter to drop to 0.6 V. That is, the large internal resistance of the battery causes a large power dissipation, which makes the transmitter stop functioning.

In this study, the authors developed a urinary incontinence sensor system [3] that combines a urine-activated battery that outputs a voltage close to 1.5 V without an output load, an intermittent power-supply circuit with an electric double-layer capacitor (EDLC), and a 300-MHz-band wireless transmitter. The power-supply circuit has a small internal resistance, which allows the circuit to drive the transmitter at 1.3 V. The system can transmit signals over a distance of 5 m when the volume of urine is 0.2 ml.

2. Urine-Activated Battery

Our urine-activated battery (Fig. 1) consists of a Zn anode, a paper suspension, a separator, and a sheet of MnO2 and carbon for the cathode. The whole assembly is sandwiched between the two parts of a metal case. The lower part of the case has holes that allow urine to enter. When a drop of urine touches the battery, it is absorbed by the paper between the MnO2 and the Zn and dissolves those chemicals, which then react to produce electricity. Equations (1) and (2) show the chemical reactions at the anode and cathode, respectively:

\[ \text{Zn} + 2\text{OH}^- \rightarrow \text{Zn(OH)}_2 + 2e^- \]  
\[ 2\text{MnO}_2 + 2\text{H}_2\text{O} + 2e^- \rightarrow 2\text{MnOOH} + 2\text{OH}^- \]

and the overall reaction is

\[ \text{Zn} + 2\text{MnO}_2 + 2\text{H}_2\text{O} \rightarrow 2\text{MnOOH} + \text{Zn(OH)}_2 \]  

The nominal voltage for this kind of zinc-carbon battery is 1.5 V. A fabricated prototype (Fig. 2 (a)) has a diameter of 2.25 cm, a thickness of 1.8 mm, and a weight of 1 g. The holes are 3.6 mm in diameter. The battery has three parts, each of which consists of three components (Fig. 2 (b)): The upper and lower parts consist of a metal case, a metal electrode, and a Ti mesh. The mesh is welded to the case to keep the electrode in place. The middle part consists of a separator, a paper suspension, and a gasket.
The change in output voltage over time (Fig. 3) was measured for four people using a volume of urine of 0.2 ml and no output load. The battery starts outputting a voltage of over 1.3 V at around 90 s.

To determine which component of urine is mainly responsible for the generation of power in the battery, we prepared artificial urine [4] from the ingredients in Table 1. For this experiment, we used a simple parallel-plate battery in a beaker and connected it to a 10-mF capacitor to enable precise measurement of the output voltage. The electrodes were Zn and MnO₂ sheets with a length of 35 mm, a width of 5 mm, and a spacing of 6 mm. A 20-mm-long section of the electrodes was immersed in the artificial urine. The output voltage characteristics (Fig. 4) were measured for artificial urine (pH = 6), and also for a 0.3% NaCl solution and a multiple electrolyte solution (NaCl + KCl + CaCl₂) but without urea or uric acid, both of which had the same pH as the artificial urine. The output voltage for the multiple electrolyte solution is larger than that for the NaCl solution because the K⁺ and Ca²⁺ cations suppress the IR drop [5] in the solution. Moreover, the curve is close to that for the artificial urine. These results show that cations (K⁺, Ca²⁺, etc.) added to a NaCl solution are the important components responsible for power generation in a urine-activated battery.

3. Intermittent-Power-Supply Scheme

A urine-activated battery has a large internal resistance because the battery uses dilute electrolyte solution. The output voltage (Fig. 5) was measured both with an output load (1 kΩ) and without. With the output load, the output voltage drops to 1 V due to the internal resistance of the battery. Since the voltage drop at the internal resistance is 0.4 V, the value of that resistance is 380 Ω. A large amount of power is dissipated there.

The large internal resistance causes the output voltage of the battery to drop when the battery is connected to the transmitter. We devised an intermittent-power-supply scheme (Fig. 6) that employs an electric double-layer capacitor (EDLC) with a small internal resistance and two switches (SW₁, SW₂). First, the battery supplies power to the EDLC through SW₁. When the EDLC is fully charged, it starts discharging at a voltage of 1.3 V. This supplies power to the output load through SW₂. Since the internal resistance of the EDLC is much smaller than that of the battery, there is only a slight drop in output voltage.

An intermittent-power-supply circuit (Fig. 7(a)) with an EDLC makes it possible to supply a voltage of over 1 V with an output load. The circuit consists of an EDLC, three pMOSFETs (M₁-M₃), and a voltage detector. Power from the battery is fed to the EDLC, which has a small internal resistance of 92 Ω, and the EDLC supplies power to the output load. pMOSFET M₁ is the switch SW₁ for charging the EDLC, and pMOSFET
Fig. 6 Concept of power supply scheme with EDLC: (a) power supply to EDLC ($V_{i\text{n}}' < 1.3 \text{ V}$) and (b) power supply from EDLC ($V_{i\text{n}}' = 1.3 \text{ V}$).

Fig. 7 Intermittent-power-supply circuit scheme: (a) intermittent-power-supply circuit and (b) voltage detector circuit.

$M_2$ is the switch $SW_2$ for supplying power to the output load. The voltage detector determines the output voltage of the circuit. The voltage detector circuit (Fig. 7(b)) consists of an input-voltage monitor, a comparator, a $V_{\text{ref}}$ generator, and an open-drain nMOSFET driver. $V_{\text{ref}}$ is set to 0.5 V; and the resistance ratio ($R_1/R_2$) is adjusted so that the input-voltage monitor outputs 0.5 V when the input voltage of the voltage detector is 0.8 V. Furthermore, when the input voltage of the voltage detector is over 0.8 V, the output voltage of the comparator becomes high. So, the output voltage of the comparator is $V_1$ and the output voltage of the voltage detector ($V_2$) is at a low level.

The operating sequence of the intermittent-power-supply circuit is as follows: First, the EDLC starts charging until the input voltage of the circuit reaches 1.3 V. At that point, the input voltage of the voltage detector is 0.8 V due to the voltage drops of the Schottky diodes ($D_1$, $D_2$). And when the input voltage of the circuit is over 1.3 V, the voltage detector outputs a low-level voltage. Then, pMOSFET $M_2$ turns on and starts to output a voltage, while pMOSFET $M_1$ stops the charging of the EDLC. We also use hysteresis control with pMOSFET $M_3$, which is connected in parallel to the Schottky diode $D_1$, so that the circuit outputs a voltage even when the input voltage falls as low as 1 V.

The measured static-output-voltage characteristics (Fig. 8) of the intermittent-power-supply circuit show that the EDLC charges until the input voltage of the circuit reaches 1.33 V. When the input voltage is over 1.33 V, the circuit outputs a voltage that is proportional to the input voltage. The circuit continues to output a voltage as the input voltage decreases down to 0.95 V.

The measured dynamic-output-voltage characteristics (Fig. 9) show that, when the voltage of the EDLC is over 1.3 V, the circuit outputs a voltage of 1.3 V and continues to output a voltage until the output voltage drops to 0.95 V. The hold time of the output voltage is 900 ms when the EDLC has a capacitance of 3 mF.

4. Self-Powered Wireless Urinary Incontinence Sensor System

Our self-powered wireless urinary incontinence sensor (Fig. 10) consists of a urine-activated battery, an intermittent-power-supply circuit with an EDLC, and a wireless transmitter. Since the transmitter operates intermittently, it has a very small power dissipation. The system uses a 315-MHz-band low-power wireless transmission scheme with a SAW oscillator.

The SAW oscillator (Fig. 11) is a Colpitts-type oscillator consisting of a SAW resonator, a bipolar transistor, and an LC circuit. The equivalent circuit of the SAW resonator contains an inductor, a capacitor, and an internal resistor connected in series; and a parasitic capacitance is connected to these elements in parallel. The output impedance of the SAW resonator is equal to the sum of the parasitic capacitance of the SAW resonator and the input impedance of the bipolar transis-
tor, which includes $C_1$, $C_2$, and $R_E$. Since it is converted into an impedance consisting of a negative resistance and a capacitive reactance connected in series, the oscillator starts oscillating when the negative resistance is larger than the internal resistance of the SAW resonator. The SAW oscillator, in which a SAW resonator directly drives the circuit, is suitable for intermittent operation because it responds quickly to the on-off control of the power supply.

A fabricated wireless transmitter (Fig. 12(a)) is 2.15 cm $\times$ 1.95 cm in size. The measured RF output spectrum (Fig. 12(b)) shows the output power of the occupied bandwidth to be about $-10$ dBm for a 50-ohm load at a supply voltage of 1.3 V. The power dissipation of the oscillator is 1.3 mW.

The measured output power of the transmitter for various supply voltages (Fig. 13) shows that it functions when the voltage is over 0.9 V. The power dissipation drops in proportion to the supply voltage as the supply voltage decreases down to 1.1 V and then drops sharply. So, we set the supply voltage of the oscillator to 1.3 V to keep the variation in output power under $\pm 14\%$ for a supply voltage variation of $\pm 10\%$.

Our urinary incontinence sensor system (Fig. 14) consists of a urine sensor and a receiver. The sensor (Fig. 15) is 4.3 cm $\times$ 4.2 cm $\times$ 1.45 cm (LWH) in size and weighs 3 g. The transmitter radiates a power of $-10$ dBm at 1.3 V, and the maximum antenna gain is $-20$ dBi. The receiver consists of an RF front-end and a digital controller. The RF front-end detects the 315-MHz carrier and outputs a binary signal. The received power of the receiver is $-85.5$ dBm at a bit-error rate (BER) of $10^{-2}$, and the antenna gain is -12 dBi. From the transmitter power, the received power, and the gains of the antennas, the free-space path loss (FSPL) was calculated to be $-43.5$ dB. The transmission range ($D$) was calculated using the following equation

$$D = \left( \frac{\lambda}{2\pi} \right)^{10 \text{FSPL}}$$

(4)

where $\lambda$ is carrier wavelength. The value was found to be 11 m.

we poured artificial urine drop by drop into four holes of urine-
activated battery by using a dropping pipette. The volume of a drop of urine was 0.2 ml. A urine sensor transmitted signals over a distance of 5 m.

5. Conclusion

A wireless self-powered urinary incontinence sensor system with a urine-activated battery has been developed. The battery both senses urine leakage and provides power. An intermittent power-supply circuit with an electric double-layer capacitor (EDLC) allows the battery to drive a wireless transmitter. This circuit and a 1-V SAW oscillator make ultralow-power wireless transmission possible. The SAW oscillator responds quickly to the on-off control of the power supply, which makes it suitable for intermittent operation. A small urine incontinence sensor system in a diaper will be useful for the care of bed-ridden elderly people in hospitals.

Acknowledgments

The authors would like to thank Dr. M. Hayashi of NTT Corporation for his support in fabricating the urine-activated battery and Prof. Y. Endo of the Shiga University of Medical Science and Prof. H. Shiraishi of Ritsumeikan University for helpful discussions regarding the composition of urine. The authors also thank Mr. N. Hama of Seiko Epson Corporation for his support in fabricating the wireless system.

References


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