Urine-Powered Wireless Urinary Tract Infection Monitoring Sensor For Smart Diaper Platform

Annual Report
(August 2015 ~ July 2016)

Project Sponsored by
Catalyst Foundation

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Summary

In the first year, we developed the first version of a urinary track infection (UTI) sensing system consisting of custom designed urine-powered battery, optical colorimetric UTI sensor, low power sensor interface circuits, Bluetooth module, and an Android app displaying the sensing results. The circuits that include DC-DC converter, LED driver, photo-diode interface, and pulse width modulator, are optimized for low power consumption. An algorithm that can compensate process variations has been developed and implemented into an Android app that calculates and displays the final sensing results.

In the second year, we presented the results of the first year work at the 2015 IEEE Biomedical Circuits and Systems (BioCAS) Conference. The work also has been invited to the 2016 International Conference of the IEEE Engineering in Medicine and Biology Society. The first version developed during the first year has four separate modules connected through wires: battery module, sensor module, sensor interface circuit module, and Bluetooth communication module. During the second year, we reduced the number of modules to two: (a) a disposable module that includes urine-powered battery, sensing unit, and sensor interface circuit, and (b) reusable Bluetooth communication module. Only the low-cost disposable module will be embedded into a diaper. The sensor interface design has been further optimized for improving power efficiency, and the battery has been redesigned to improve its flexibility, so it can be easily embedded in a diaper without causing any irritation. We are also conducting test with natural human urine, instead of using synthetic urine. To further reduce power and size, we are designing a custom integrated circuit (IC) that includes DC-DC converter, sensor interface and wireless circuit. A journal paper including a part of the second year results has been submitted to the IEEE Transactions on Biomedical Circuits and Systems.

In the following years, we will focus on power and size reduction by utilizing custom IC, and reliability and reproducibility test utilizing human urine samples.

[Publications]


I. Introduction

Urinary tract infection (UTI) is the second most common infection in the body accounting for more than 7 million office visits and 100,000 hospitalizations per year. Although mostly uncomplicated and easily treatable using antibiotics, if not identified and treated early, UTI can be a major source of morbidity and mortality such as ascending infection, loss of kidney function, and sepsis. The problem is particularly significant in geriatric patients, in particular, those suffering from neurodegenerative diseases such as dementia who require long-term care in assisted living facilities. Also, for infant population. UTI is one of the most common causes of renal damage, worsened by a delay in diagnosis. Many of the disabled elderly, young children, and all infants using diapers are not capable of understanding the symptoms of a UTI, and many who are capable have difficulties in communicating the symptoms to caregivers. Consequently, early identification and treatment of UTIs is vital to the prevention of major sequelae or death.

Urine culture test is the most conventional UTI diagnose. One significant disadvantage of the current microbiology method is a lapse of 2 to 3 days between specimen collection and final diagnose. As a result, dipsticks have been widely used for less accurate but quick detection of UTIs. However, in addition to potential privacy issues, collecting urine samples from infants and disabled elderly for dipstick test is not trivial. In this regard, a better method for early detection and screening of UTIs that can alert caregivers with minimal efforts is being sought after.

During the first year of the project, we investigated a diaper-embedded, wireless, self-powered, and autonomous UTI monitoring sensor module that allows effortless early detection and screening of UTIs. Figure 1 shows a conceptual diagram representing the sensor module and system. As UTI dipsticks detect the level of nitrite, a surrogate of UTIs, the sensor module measures the concentration of nitrite in the urine, and transmits the data via a Bluetooth low energy (BLE) module to a nearby BLE capable mobile device. The sensor module utilizes a flexible substrate that can be embedded in a diaper during the manufacturing process or attached as an added unit to a commercially available diaper. It is highly desirable that the disposable sensor module embedded in a diaper is self-powered because of cost and environmental issues. Therefore, we utilized a urine-activated battery for self-powered autonomous operations. Because of the limited amount of energy that the urine-powered battery can provide, a power efficient yet accurate sensor interface circuit is crucial for successful implementation, and thus we developed a low-power sensor interface design utilizing pulse width modulation.

![Figure 1. Diaper-embedded UTI monitoring system](image)

II. Review of the principle and design

2.1 System architecture

As shown in Figure 2, the diaper-embedded UTI monitoring sensor module consists of a paper-based colorimetric nitrite sensor, four urine-powered batteries connected in parallel, a boost DC-DC converter, a low-power sensor interface utilizing pulse width modulation, and a BLE module for wireless transmission. Once the
urine reaches the batteries, they are activated and start to provide power to the rest of the sensor module, hence waking up the whole sensor module. This urination-event driven automatic wakeup eliminates the need for periodic wakeup for checking urination-event, which is required for a system that relies on a continuously powered battery. The output voltage of the urine-powered battery ranges from 0.3V to 0.9V, and the boost DC-DC converter increases the voltage to a regulated voltage of 2.0V. The level of power provided by the urine-powered battery varies from 3mW to 9mW, and the power conversion efficiency of the DC-DC converter falls in the range from 50% to 60%. Then, the boosted voltage powers up the sensor consisting of a light emitting diode (LED), a urine-absorbing strip, photodiodes, the sensor interface, and the BLE module. When the urine-absorbing strip absorbs the urine, the color of the strip changes from white to pink. As in UTI dipstick, the color density of the strip is proportional to the nitrite concentration in the urine and affects the amount of the light that reaches the photodiode, and consequently determines its photocurrent. The sensor interface converts the photocurrent into a pulse width modulated (PWM) signal. The BLE module has a built-in counter operating at 40kHz, which transforms the PWM signal into a digital signal. This architecture utilizing pulse-width-to-digital conversion eliminates the need for an ADC, which significantly improves the power efficiency and reduces the complexity of the system. The BLE module transmits the data to a nearby BLE capable mobile device of a caregiver, which is presumably connected to a healthcare network.

Fig. 2. A block diagram of the diaper-embedded UTI monitoring sensor module
(a) Urine-activated batteries (b) Boost DC-DC converter (c) Colorimetric nitrite sensor
(d) Sensor interface (e) BLE module

2.2 Sensor interface with pulse width modulation

In the colorimetric nitrite sensor, the intensity of the light reaching the photodiode after passing through the urine-absorbing strip carries the information of the nitrite concentration in the urine, indicating the level of UTIs. To interpret the light intensity signal into electrical signal, typically a transimpedance amplifier (TIA) is used to turn the photodiode current into voltage, which is further encoded into digital data by an analog-to-digital converter (ADC). This conventional method increases the complexity and the power consumption of the interface circuit. Due to the limited amount of energy available from the urine-powered battery, a power efficient sensor interface is highly desirable for reliable and self-powered operation of the sensor module. To address these issues, we developed a semi-digital PWM-based method in this work, where the pulse width of the binary output signal is inversely proportional to the photodiode current, and a counter translates the pulse width into digital information. This PWM-based method allows eliminating the complex and potentially power-
intensive TIA and ADC. It also provides a wide dynamic range because the data is represented in the time domain as the pulse width rather than the voltage or current domain where its dynamic range is typically limited by the supply voltage.

For the pulse width modulation operation, two reverse biased photodiodes are connected in series, and the shared node, $V_O$ node, between the two photodiodes serves as a high impedance capacitive node charged by the photocurrent. The parasitic capacitors, $C_1$ and $C_2$, of the two photodiodes serve as a charging capacitor. We note that the shielded dummy photodiode at the bottom not only contributes to the capacitive impedance node required for charging, but also compensates the leakage current. Because the two photodiodes have almost the same leakage currents and the leakage current is insensitive to voltage bias conditions, the balanced leakage currents of the photodiodes do not charge the output node, $V_O$. The PWM modulator consists of two inverters, three NAND gates, and an analog switch. Figure 3 shows the timing diagram of its operation. At the rising edge of the CLK signal provided by the BLE module, the active-high RESET signal goes LOW, turning the switch SW off and initiating the charging. As the photocurrent $I_{PD}$ charges the capacitive node, the output voltage $V_O$ increases. Once the voltage $V_O$ reaches $V_{TH}$, the threshold voltage of the inverter, the inverter output pushes the RESET signal to HIGH, resetting $V_O$ to the ground. We use the inverse of RESET as PWM signal. The relation between the photocurrent $I_{PD}$ and the pulse width $PW$ can be described as follows:

$$PW = \frac{\alpha}{I_{PD}}$$

where $\alpha = (C_1 + C_2) \cdot V_{TH}$.

### 2.3 Calibration for process variation

There exists a significant level of process variations in the LED, photodiodes, and the urine-absorbing strip. The large variation brought by the consisting elements would make a reliable operation of the sensor module almost impossible, so an effective on-line calibration method is essential.

We note that the process variations would affect the output pulse width readings for the dry urine-absorbing strip (before absorbing) and for the wet urine-absorbing strip (after absorbing) in the same manner. Consequently, the difference between the pulse width for the dry urine-absorbing strip ($PW_1$) and the pulse width for the wet urine-absorbing strip ($PW_2$) would suffer from process variations much less. To measure the difference, we use a time lapse reading technique. The sensor module is designed in such a way that the urine reaches the urine-powered battery first, and the urine-absorbing strip later. As soon as the urine-powered battery becomes active, the module measures the pulse width ($PW_1$) while the strip is dry. The following readings use $PW_1$ as a reference, and the sensor module registers the difference, $PW_D = PW_1 - PW_2$, rather than $PW_2$. Although this time lapse based differential reading reduces the effect of process variations significantly, the differential reading, $PW_1 - PW_2$, still suffers from process variations.

To further reject the adverse effects of process variations, we use a calibration technique utilizing a known reference. In this technique, we use one arbitrarily selected sensor module as a known universal reference. First, we measure the nitrite concentration versus the differential pulse width curve using the known reference module.
as shown in Figure 4.

We assume that for each component the deviation of the performance of a building component in an end-user module from that of the component in the reference module is linear. For example, the light intensity from an LED in an end-user module will be different from that in the reference module, resulting in different output pulse widths. We assume that the difference between the two can be calibrated by multiplying a constant number. Assuming it is true for other components in the module, the measured output pulse width of an end-user sensor module can be annotated back to the pulse width of the reference module by multiplying a constant $K$, as shown in Figure 4. Because each end-user module will have a different value of the constant $K$, each sensor module should be able to find its own value of $K$ unless the constant value is measured for each module during the production stage. In the developed system, each sensor module finds the value of $K$ autonomously utilizing the dry condition measurement. For the time lapse based differential measurement, the sensor module first measures the pulse width $PW_{1,Dry}$ when the urine-absorbing strip is dry. Because the constant $K$ should not be affected by the nitrite concentration in the urine, we can calculate the constant $K$ using the measured $PW_{1,Dry}$ and the known value of $PW_{1,Dry,REF}$ from the known reference.

$$K = \frac{PW_{1,Dry,REF}}{PW_{1,Dry}}$$

(2)

The calibrated pulse width $PW_{CAL}$ can be obtained by multiplying the calculated constant $K$ to the measured differential pulse width:

$$PW_{CAL} = K \cdot PW_1 - K \cdot PW_2 = K \cdot PW_D$$

(3)

III. Experimental results

The measurement setup used for the sensor module is shown in Fig. 5. The boost DC-DC converter and the PWM sensor interface are mounted on a custom printed circuit board (PCB). During measurements, the colorimetric nitrite sensor is placed in a box to make sure that the active photodiode only responds to the LED light, but not to the ambient light. An LED with the peak emission wavelength of 572nm is chosen because of its high sensitivity to pink color. Two photodiodes with the peak sensitivity wavelength of 540nm are used and the LED bias current is set to 0.5mA while the PWM sensor interface draws 0.08mA from the regulated 2V supply. The microcontroller in the BLE module (RFD22102 from RFDuino) provides the CLK signal and converts the pulse width output of the PWM sensor interface into digital data using the built-in counter that is transmitted to a paired mobile device (an Android tablet). The BLE module draws 4mA in transmission mode.

Synthetic urine samples with nitrite concentrations of 0mg/L, 4mg/L, 6mg/L, 8mg/L, and 10mg/L are used for the measurements. The PWM signals for the dry reagent strip and for the wet reagent strip are measured as shown in Fig. 6. Fig. 7 shows $T_{PW1}$, $T_{PW2}$, and the differential pulse width displayed on the mobile device at real-time. Fig. 8 illustrates the differential pulse width versus nitrite concentrations curves for ideal case simulation.
and measurements before and after the calibration. For the measured differential pulse width $T_{DPW}$ before the calibration, the maximum error from the ideal case simulation reaches up to 33%, while the $T_{DPW}$ after the calibration shows a maximum error of 3.5%, which corresponds to almost 10 folds of improvement. The sensor module achieves a sensitivity of 1.35 ms/(mg/L) and a detection limit of 4 mg/L for nitrite. Because nitrite is never found naturally in urine, and many species of gram-negative bacteria convert nitrate to nitrite, the $T_{DPW}$ of urine from a person without UTI will show 40ms. The $T_{DPW}$ of the urine of a person with UTI will be smaller than 40ms, and the difference will increase with increasing amount of nitrite in the urine. Fig. 8 (b) shows a reference color chart of typical urine dipsticks designed for nitrite detection. To the untrained eyes, a quantitative analysis on nitrite concentration, beyond the decision on positive or negative, is not feasible for dipsticks. The comparison demonstrates the effectiveness of the proposed autonomous sensing utilizing the developed sensor module.

Fig. 6. Measured PWM signal (a) Pulse width for the dry reagent strip: (i) Clock signal, (ii) Output $V_{PWM}$ signal, and (b) Pulse width for the wet reagent strip: (i) Clock signal, (ii) Output $V_{PWM}$ signal.

Fig. 7. Displayed result on the mobile device.
IV. Conclusion

We developed a diaper-embedded UTI monitoring sensor module self-powered by a urine-powered battery. The sensor module can detect UTIs autonomously and transmit the data to a caregiver’s mobile device that is presumably connected to a healthcare network, allowing a reliable early detection and screening of UTIs with minimal efforts. The sensor module consists of a paper-based colorimetric nitrite sensor, a urine-powered battery, a boost converter, a low-power PWM sensor interface, and a BLE module. The urination-event driven wakeup method allows a simple yet efficient operation of the system relying on limited amount of energy from the urine-powered battery. The sensor interface applying PWM method significantly reduces the circuit complexity and the power consumption. The time-lapse-based differential reading scheme and the linear calibration utilizing a known reference significantly reduce the adverse effects of process variations in the components without increasing the system complexity. With the calibration, the maximum error in nitrite concentration estimation has reduced from 33% to 3.5%. The sensor module achieves a sensitivity of 1.35 ms/(mg/L) and a detection limit of 4 mg/L for nitrite.

V. Future Work

In the following years, we will focus on power and size reduction by utilizing custom IC, and reliability and reproducibility test utilizing human urine samples. We are currently designing a custom IC in TSMC 130nm technology. We are also working with a urologist at the Indiana University Hospital for human urine sample test.