A Wireless Bladder Volume Monitoring System Using a Flexible Capacitance-based Sensor

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Abstract — A wireless sensor system has been prototyped to monitor bladder volume in a small animal. The capacitance-based sensor was laser-micromachined from a polyimide film to form an interdigitated finger structure, followed by a sputtering process to deposit aluminum thin film. The entire sensor was encapsulated with biocompatible and elastic polydimethylsiloxane (PDMS). A passive telemetry platform based on inductive coupling was realized to employ the sensor in vivo and a commercial wireless module was used for data communication in air. The entire system consists of a sensor implant, an external wearable unit and a base station. The system was tested with a small rodent model demonstrating the feasibility and identifying practical issues in implementation.

Index Terms — Urinary incontinence, bladder volume monitoring, capacitive-based sensor, passive wireless system.

I. INTRODUCTION

Urinary incontinence (UI) is the involuntary loss of urine due to lack of bladder control. In 2000, the total annual costs for UI have exceeded $19.5 billion [1]. Besides discomfort and inconvenience, UI may lead to serious problems such as kidney failure. In some cases, urinary bladder dysfunction cuts off the sensorial feedback to the central nervous system, thus patients are incapable of knowing their bladder conditions or when is the time to consider voiding their bladders. Urine may flow back into kidneys from a full bladder, causing damages which result in hemodialysis [1-2].

Several methods have been used to detect bladder volume with invasive sensors or noninvasive measurements [2-3]. Catheterization and bioelectric impedance probing cause potential side effects, ultrasound measurement limits patients’ mobility, whilst electroneurogram requires complicated data processing [2] [4]. Strain sensors have been developed to measure bladder volume by several research groups. However, the approach of invasively wrapping polypyrrole around a bladder could not be easily implemented while the carbon film sensor gave a low sensing range of 2% which would not be sufficient for the bladder volume changes [2] [3].

In our previous work [4], we have developed a strain sensor integrated with a wireless system to monitor continuously the volume of a human bladder model. The strain sensor was based on a variable capacitance employing a metal interdigitated finger structure encapsulated in biocompatible polymers. The system consists of three components: an implanted transponder; a wearable unit that transfers electromagnetic energy to the implant, receives the sensor signals from the implant and processes the signals in a microprocessor before sending out wirelessly; and a base station that receives and records the wireless data. The system was verified with a bladder phantom. In order to further validate the performance of such an implant system, trials on an animal model are needed prior to human uses. There is also a need for monitoring bladders in animals for urinary incontinence drug development. Ideally, the animal models should mimic human conditions. However, large animal models such as pigs are expensive due to surgery and care costs. Therefore, small animal models are proposed.

In this paper, the sensor material was changed to flexible polyimide to enhance sensor elasticity and longevity. The device size was reduced so that it could be used in rodents. The attachment procedures were also modified to allow sensors working with small bladders. Experimental data were recorded wirelessly with rodents under anesthesia.

II. DESIGN AND IMPLEMENTATION

A. Sensor Fabrication

Interdigitated capacitor (IDC) structures were created with a 125-μm thick polyimide film using a laser micromachining system. The finger length was 500 μm and the separation gaps between fingers were 5 μm. In order to reduce the size of the sensor, the number of fingers was chosen to be 40 instead of 60 in our previous work [4]. Bridges were designed to hold the two parts of the IDC. The IDC structure cutouts were then coated with 4-μm thick Al with sputtering for three hours. The long-period sputtering was to ensure the sidewalls of fingers were also metalized. The IDCs were then attached onto a thin, half-cured PDMS (Sylgard 184) layer which was spun on a glass slide. After the PDMS was cured, wire connections were made with silver epoxy. The bridges were cut off, followed by casting a top layer of PDMS to encapsulate the entire sensor. The PDMS layer with sensor embedded was then peeled off and the entire sensor was tailored to remove excessive PDMS. Fig. 1(a) shows
The frequency generated can be realized by inductive coupling between two coil realizations. (6) The capacitance decreased as the bending angle increased, and varied with strain. (7) The results show that the strain-generating sensor is effective for monitoring the bending of PDMS sets.

The capacitance of the IDC can be expressed as

$$C_{IDC} \approx \frac{\varepsilon (L - d' - x) \alpha \beta (N - 1)}{d} + \frac{\varepsilon \omega a' b' N}{d' + x}$$

where $\varepsilon$ is the effective permittivity of the PDMS, $L$ is the finger length, $x$ is the finger movement due to strain, $t$ is the thickness of the polyimide, $d$ is the separation gap, $d'$ is the separation gap at the tips of fingers, $w$ is the width of fingers, $N$ is the number of the fingers, $\alpha$ and $\alpha'$ are the terms related to the fringe effects of electric fields, $\beta$ and $\beta'$ represent the effects from finger sidewalls which are not perfectly in parallel due to the finite laser beam spot and repeated laser writing. As the polymer is stretched, the two parts of the IDC are pulled away from each other and the overlap length between the fingers decreases, resulting in a decrease in capacitance. When the IDC is bent instead of stretched, the fingers are tilted and the capacitance decreases due to electric field distribution changes. In practical uses, the variation of capacitance is likely contributed by both effects.

### B. System design

Fig. 2(a) shows the conceptual design of the system we envision to be used in patients [4]. The radio frequency (RF) powering from the wearable unit to the implant was realized by inductive coupling between two coils at resonance. The LC circuits in both sides were tuned at 1.3 MHz. The external unit was designed to be embedded in a belt so that the patient can wear around the abdomen. The communication between the wearable unit and the base station was based on two eZ430RF2500 (Texas Instruments) modules [4-5]. The wireless communication was for the future purpose of continuous monitoring a group of freely-moving animals. In practical applications, the wireless communication will be used to care paralyzed patients and patients who have cognition or mobility issues.

In the implant, a relaxation oscillator circuit was used to convert the IDC capacitance changes to frequency variations (Fig. 2(b)). The frequency generated can be calculated as [4]

$$f = \frac{1}{2 \ln(3) R C_{IDC}}$$

where $R_d$ is 3 M$\Omega$. With the range of $C_{IDC}$ in 8–10 pF, the modulation frequency is in the range of 15–18 kHz. The component values were modified from the ones in [4] to adapt to the smaller sensor.

### III. EXPERIMENTS AND RESULTS

Male Sprague–Dawley rats (~400 g) were used for experiments. The rats were anesthetized with sodium pentobarbital (50 mg/kg) maintaining normal physiological conditions. The bladder was exposed for sensor attachment.

With rat’s smaller bladder, the sensor attachment method using medical glue for the phantom in [4] was not suitable. The glue locally hardened the bladder tissue affecting strain transduction. The entire sensor was therefore tailored into a long strip (Fig. 1(c)) so that it could be wrapped around the bladder. The sensor strip was clamped in the ends to secure contact. As the bladder expanded, the sensor would be stretched and bent causing a shift in the output frequency.

The relationship between sole stretching and varied capacitance has been investigated with a cantilever setup as reported in [4]. An apparatus was developed to precisely bend the sensor while the capacitance was measured. Fig. 3 shows the setup configuration and measured result. The capacitance decreased as the bending angle increased.
In this preliminary work, the implant circuit on a printed circuit board was placed outside the body and the implant coil was at a fixed distance of 6 cm from the external unit coil. The bladder was placed near the incision in order to visually observe the volume and sensor deformation. A syringe pump was set up to manually fill the rat bladder with saline through the urethra. A clamp was used to block the urethra so that the bladder volume could be maintained, as shown in Fig. 4(a). The initial volume of liquid inside the bladder was set at 1 ml. An incremental volume of 100 µl was gradually injected. The output frequency from the implant circuit to the computer via wireless communication was recorded continuously.

As saline was injected into the bladder, the output frequency decreased indicating increases of IDC capacitance. The capacitance was low at the initial point due to the high curvature of a small bladder. As the bladder expanded, the bending curvature was reduced causing an increase in capacitance. The sensor was also stretched but the change in capacitance was dominated by the bending effect. Fig. 4(b) shows the experimental result. The bending angle from the initial point was calculated as 8° with values in Eq. 2 and Fig. 3.

The modulated frequency presented a monotonic relationship with the incremental bladder volume. The relative shift of frequency could distinctly indicate the bladder volume change. From our past experience, the 6-cm spacing between coils in air indicated that the RF powering would be sufficient for sensor signal transduction through fatty tissues with a 3-cm thickness.

In this work, we aim to develop a miniature, passive, implantable sensor as well as a wireless monitoring system for the uses in rodent models for experiments. The purposes are to validate the sensor functionality and investigate practical issues in future animal experiments for therapeutic drug and stimulator implant development. In future applications for human, further investigation is needed to find proper implantation locations and appropriate attachment methods.

IV. CONCLUSION

A bladder volume monitoring system for small animals using a batteryless wireless transponder platform integrated with a flexible capacitive sensor has been successfully demonstrated. The system was capable of detecting small changes of volume in a rat’s bladder and recording the data remotely via wireless communication. The preliminary results show promise for implantation and remote monitoring in freely-moving rodents.

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