Annals of Biomedical Engineering, Vol. 49, No. 1, January 2021 (© 2020) pp. 149–161
https://doi.org/10.1007/s10439-020-02528-7

BMES BIOMEDICAL ENGINEERING SOCIETY

Original Article

An Instrumented Urethral Catheter with a Distributed Array of Iontronic Force Sensors

Ye Zhang,[1] Mahdi Ahmadi,[1] Gerald Timm,[2] Serdar Sezen,[3] and Rajesh Rajamani[1]

[1]Department of Mechanical Engineering, University of Minnesota, Minneapolis, USA; [2]Department of Urology, University of Minnesota, Minneapolis, MN 55455, USA; and [3]Department of Mechanical and Manufacturing Engineering, St. Cloud State University, St. Cloud, USA

(Received 19 November 2019; accepted 29 April 2020; published online 6 May 2020)

Associate Editor Tingrui Pan oversaw the review of this article.

Abstract—This paper develops a novel instrumented urethral catheter with an array of force sensors for measuring the distributed pressure in a human urethra. The catheter and integrated portions of the force sensors are fabricated by the use of 3D printing using a combination of both soft and hard polymer substrates. Other portions of the force sensors consisting of electrodes and electrolytes are fabricated separately and assembled on top of the 3D-printed catheter to create a soft flexible device. The force sensors use a novel supercapacitive (iontronic) sensing mechanism in which the contact area between a pair of electrodes and a paper-based electrolyte changes in response to force. This provides a highly sensitive measure of force that is immune to parasitic noise from liquids. The developed catheter is tested using a force calibration test rig, a cuff-based pressure application device, an extracted bladder and urethra from a sheep and by dipping inside a beaker of water. The force sensors are found to have a sensitivity of 30–50 nF/N, which is 1000 times larger than that of traditional capacitive force sensors. They exhibit negligible capacitance change when dipped completely in water. The pressure cuff tests and the extracted sheep tissue tests also verify the ability of the sensor array to work reliably in providing distributed force measurements. The developed catheter could help diagnose ailments related to urinary incontinence and inadequate urethral closure pressure.

Keywords—Instrumented catheter, Urethral sensors, Catheter pressure sensors, *In vivo* force sensors, Iontronic sensors, Parasitic capacitance, Supercapacitive sensors.

Address correspondence to Rajesh Rajamani, Department of Mechanical Engineering, University of Minnesota, Minneapolis, USA. Electronic mail: rajamani@me.umn.edu

INTRODUCTION

Urethral Catheter

Catheters are used for both diagnostics and therapy in many medical applications, including cardiovascular, gastro-abdominal and urinary tract diseases.[11] Physicians and surgeons are well versed with using catheters, moving them through natural passageways in the body and performing diagnosis as well as delivering therapeutic devices to address medical problems that include blockages and failed valves.[9,16,26] This paper develops a unique catheter with an array of distributed force sensors that can be used to measure closure pressure in the urethra or other body passageways.

Urethral catheters can be useful in diagnosing the cause of urinary incontinence (UI). UI is "the complaint of any involuntary leakage of urine".[1] Urinary incontinence is not a life-threatening or dangerous condition, but it is socially embarrassing and may cause withdrawal from social situations and reduce quality of life.[22] An estimated 80% of people affected are women. Urinary incontinence is believed to affect at least 13 million people in the United States, and this number is expected to increase sharply with the aging of the baby boomers.[7,10,13,30,31,34] Globally, up to 35% of population over the age of 60 years is estimated to be incontinent.[12] It is estimated more than 50% of nursing facility admissions are related to incontinence.[29]

Figure 1 shows a human bladder and urethra in which a catheter is inserted through the urethra such that the tip of the catheter is inside the bladder. If such

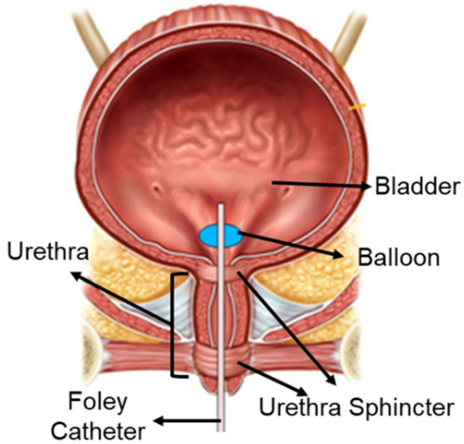

FIGURE 1. Foley catheter inserted into urethra during urodynamic test.

a catheter could be equipped with a pressure sensor at the tip and additional distributed sensors which can be located inside the urethra, then bladder pressure and distributed urethral pressure could be simultaneously measured. This would help diagnose the cause of UI and to differentiate between the bladder and the urethra as the source of UI problems. For example, changes in bladder and urethral pressure could be measured during provocative maneuvers such as coughing, laughing, pressing on the abdomen and val salva. This could help identify whether an overactive bladder, a collapsed bladder, an overactive urethral sphincter or inadequate urethral closure pressure is the source of UI in an individual patient.[25,32]

This manuscript develops an instrumented urethral catheter which has five distributed force sensors on the catheter. The pressure distribution inside the urethra is not uniform. The closure pressure is low at the two ends of the urethra and is maximum at a point approximately in the middle. Since anatomical size of a human subject varies, it is necessary to measure pressure at multiple points in the urethra, and not just at one point. Hence, a catheter with a distributed set of 5 sensors is developed in this manuscript. The force sensing units are modular and individually fabricated so that their locations on the catheter could be adjusted as desired, for example placing one at the tip of the catheter and others at a location corresponding to the urethra. The manuscript utilizes a novel sensing mechanism consisting of iontronic force transducers. In particular, this manuscript utilizes supercapacitors which employ a unique paper-based solid-state electrolyte.

Iontronic Sensing

Supercapacitors, also sometimes called as electrochemical capacitors or ultracapacitors, consist of two electrodes and an electrolyte.[5] A supercapacitor is governed by the same fundamental equation as a traditional parallel plate capacitor in which capacitance (C) can be described by

$$C = \frac{\varepsilon A}{d} \tag{1}$$

where A is the geometric surface area of the electrode; ε is the relative permittivity of the dielectric material; and d is the distance between two oppositely biased parallel plate electrodes. However, the distance d is extremely small in the case of a supercapacitor (of the order of the distance between ions at an electrode-electrolyte interface layer, a few angstroms). In use of a supercapacitor as a force sensor, a change in force is translated into a change in contact area between the electrodes and the electrolyte, and consequently results in a change in capacitance.[36] Supercapacitive sensors, also known as iontronic sensors, have much higher sensitivity compared to conventional capacitive sensors.[35]

Previous research on supercapacitive force sensors employed liquid-state electrolytes.[19,21] An example of such a sensor is shown in Fig. 2a with an electrolyte droplet being confined between two parallel electrodes.[21] This sensor provides high sensitivity and resolution. However, developing a miniaturized version of this sensor requires addressing some challenges.

Some inherent disadvantages of the liquid electrolyte sensors are:

(1) The sensors cannot be easily miniaturized to create micro-sensors, since it is difficult to create size-controlled micron-sized droplets or liquid pool and to trap the liquid inside a sealed sensor;
(2) The need for a hydrophobic coating on the surface of the electrodes reduces the capacitance, because it increases the distance between the electrolyte and the electrode;
(3) The high cost of the hydrophobic coating and the large size of the sensor pose problems in creating a sheet of such sensors for measuring distributed forces (e.g. forces from the foot of a patient as he/she walks);
(4) The effect of gravity on the liquid may limit their application in systems involving non-planar motion.

Therefore, a solid-state electrolyte is preferable to replace a liquid-state electrolyte. The deformation of the electrolytes in response to an applied force and the resulting change in its contact area with the electrodes can then be used to sense the applied force, as shown in Fig. 2b. Several iontronic pressure sensors have been reported recently. Nie *et al*[20] developed a thin-film

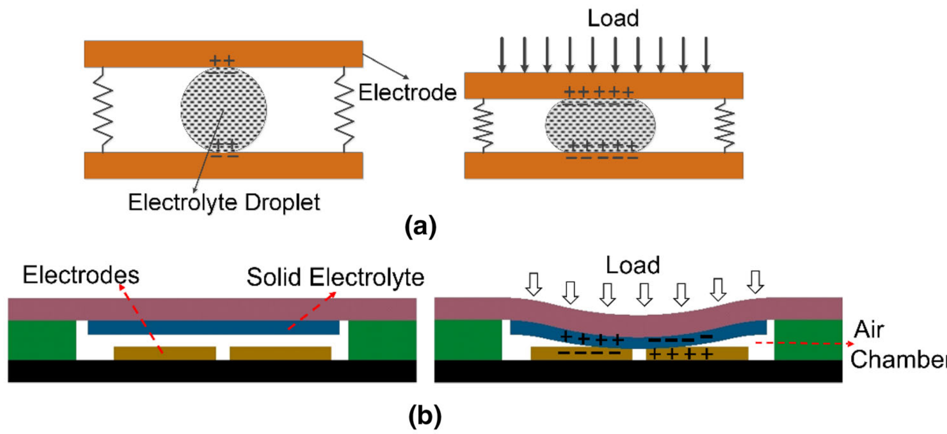

FIGURE 2. (a) Sensing mechanism of a droplet-based iontronic sensor, and (b) Iontronic sensors with solid electrolyte before and after loading.

electrolyte based on a hydrogel composite, which was incorporated in a iontronic sensor. Jin *et al* used an iontronic elastomer to develop an ultrasensitive mechanotransducer skin.[14] Li *et al* developed a wearable sensor utilizing iontronic nanofibers.[15] Zhu *et al* developed a skin-interfaced pressure-sensing architecture based on iontronic sensing.[38]

In another related work, a novel paper-based electrolyte was used for iontronic sensors made by introducing active materials into the porous matrix of paper and functionalizing the entire thickness of the paper.[36] This paper-based solid electrolyte provides enhanced mechanical elasticity properties, which enables the fabrication of iontronic sensors of high sensitivity. Further, using paper as a substrate provides easy configurability for fabricating the sensor. In this paper, the urethral instrumented catheter is developed as an interesting practical biomedical application of the paper-based solid electrolyte.

Paper Objectives

In this manuscript, an instrumented urethral catheter based on iontronic sensors is developed to measure force distribution in the urethra during clinical diagnosis in patients with UI. The developed catheter will provide simultaneous measurement of pressure at multiple locations in the urethra with one static device. The catheter is highly soft, flexible, and the sensors are able to operate in a body fluid environment. The performance of the catheter is evaluated in this manuscript by a load cell test, a pressure cuff test, an inside-water test and a test inside an extracted sheep bladder. The tests are designed to show that:

(1) The urethral catheter based on iontronic sensing has high sensitivity;

(2) The urethral catheter is immune from parasitic noise, which makes it appropriate for *in vivo* use.

The outline of the rest of this manuscript is as follows: In the next section, the design and fabrication of the catheter are presented. Subsequently, various experimental methods are used to evaluate the performance of the sensor in the following section. Results are presented in the fourth section. Discussion of the results is presented in the last section.

MATERIALS AND METHODS

Design of the Catheter

Various sensing mechanisms have been explored for sensing pressure inside the human body using a catheter. Due to the ease of miniaturization of silicon-based sensors, piezoresistive sensors have been widely used for pressure sensing in *in vivo* applications.[8,18,33] However, piezoresistive sensors need compensation for drifting.[23] Optical sensors have also been developed for *in vivo* pressure measurement, through measuring changes in intensity of the reflected light.[6,27] However, the complexity of this sensor limits its application for *in vivo* use[24]. Due to low temperature and pressure hysteresis, low power consumption and easy fabrication, capacitive pressure sensing has also been explored for use on a catheter.[2–4] However, for *in vivo* use, capacitive sensors have an inherent problem with parasitic noise in liquid environments.[2] In this manuscript, iontronic sensors are explored for use on the urethral catheter, and have features of high sensitivity, negligible parasitic noise and ease of fabrication.

The overall strategy is to fabricate a 7 Fr. (2.33 mm) catheter with five distributed force sensors based on iontronic sensing along the length of the catheter. This instrumented catheter consists of a catheter body, five iontronic sensors, and the associated electronics. Figure 3a shows the side view of the five sensors on the catheter, explaining the iontronic working principle. The bottom electrodes are patterned on a soft substrate as shown in the figure, including five separate electrodes and one additional common electrode. The electrolyte is cured on the surface of a soft deformable arch structure of a 3D printed chamber, which is assembled over each electrode. The load applied from the top of the chamber will produce a deformation in the deformable arch, which brings the electrolyte down to touch with the bottom electrodes, and consequently creates a capacitance change.

The catheter body needs to be flexible enough for the ease of insertion into the urethra. In this manuscript, 3D printing technology is used to fabricate the catheter body using a combination of both a hard material and a soft material. 3D printing is an additive manufacturing technology in which material is fused under computer control to create a three-dimensional object, with material being added together, typically layer by layer.[28] By adjusting the ratio of the soft and hard materials, the mechanical property of the catheter body can be adjusted. The sensors and the electronics are assembled on the 3D-printed catheter body with a diameter of 7 Fr.

The soft substrate with electrodes is assembled on a flat portion of the catheter body as shown in Fig. 3b. The chamber which hosts the electrolyte is also printed with a combination of both hard and soft materials (Fig. 3c) to achieve the best mechanical performance. The side wall and internal arch structure of the part is made of soft material that will deform upon loading. On the top of the soft part, a hard bump is incorporated for the purpose of better receiving the load applied and transferring it to the soft body to thus create deformation under loads. Two separators are designed as supports at the two ends of each chamber. In addition, a hard skeleton is designed and embedded inside the soft body for recovering quickly after a load is released.

The solid-state electrolyte is cured on the surface of the soft arch structure and contacts the bottom electrodes with increasing contact area when load is applied. This electrolyte is developed by introducing active materials into the porous matrix of a paper substrate. The active materials are a mixture of an ionic liquid and a photo-curable polymer. By brushing this ionic mixture on to filter paper, and then exposing the combination to UV light for a short time duration, a paper-based solid electrolyte is obtained. It has excellent mechanical properties with high flexibility and toughness and excellent electrical properties, easy fabrication process and low cost. Iontronic sensors using the paper-based solid electrolyte as sensing element provide high sensitivity, high sensing range, and good ion mobility.[36]

Fabrication of the Instrumented Catheter

The catheter body is made with 3D printing technology using both hard and soft materials as shown in Fig. 4a. The as-fabricated catheter body has a shore value of 70 A. The diameter of the catheter body is 7 Fr. Near the tip of the catheter, a flat stage is printed to host the electrode substrate. Another stage with grooves is designed near the other end of the catheter body for the location of the connectors and wires.

The bottom metallic electrodes are fabricated on a flexible substrate, using MEMS fabrication techniques, as shown in Fig. 4b. First, a thin layer of copper (30 nm) is sputtered on a polyimide (PI) film before a gold layer (20 nm) is sputtered. PI film is lightweight, flexible, resistant to heat and chemicals, and thus widely used as an electronic circuit substrate. The top surface of gold is the effective electrode material for sensing, which contacts the solid electrolyte upon loading. The size of each electrode is 0.4 mm $\times$ 8 mm. Then, the five electrodes and one common ground electrode are patterned using photolithography and wet etching. The as-fabricated electrodes on the soft and flexible PI film are transferred onto the catheter body. Figure 4c shows a photograph of the electrodes on PI substrate on the catheter body while it is being bent.

The 3D printed chamber with electrolyte is shown in Fig. 4d. The part is made using multi-material 3D printing technology, combining materials of different properties in a single part. The structure contains a soft body, a hard bump, two hard separators and a hard skeleton. The bump is printed for the purpose of better receiving force applied on top, while the hard skeleton is embedded inside the soft body and designed to improve the recovery after force is released. The chamber was printed as a whole using multiple materials. The Stratasys J750 PolyJet 3D Printer was utilized.

The fabrication of the paper-based solid electrolyte is shown in Fig. 4e. The filter paper is pre-shaped into an arch and transferred to the 3D printed chamber. An ionic liquid, a prepolymer solution, and a photo initiator are mixed in the ratio of 5:4:1 by weight by sonicating. The mixed gel is then brushed on to the filter paper, that functionalizes the entire thickness of the paper. After that, the combination was put under UV light (~ 325 nm, 36 W) in a glovebox (inert environment filled with nitrogen) at room temperature for 1 min. A clear film (150 μm) sticking to the 3D printed chamber is obtained. It should be noted that the ratio of the components can be changed to achieve solid electrolytes with different mechanical properties. According to our previous research,[36] this novel paper-

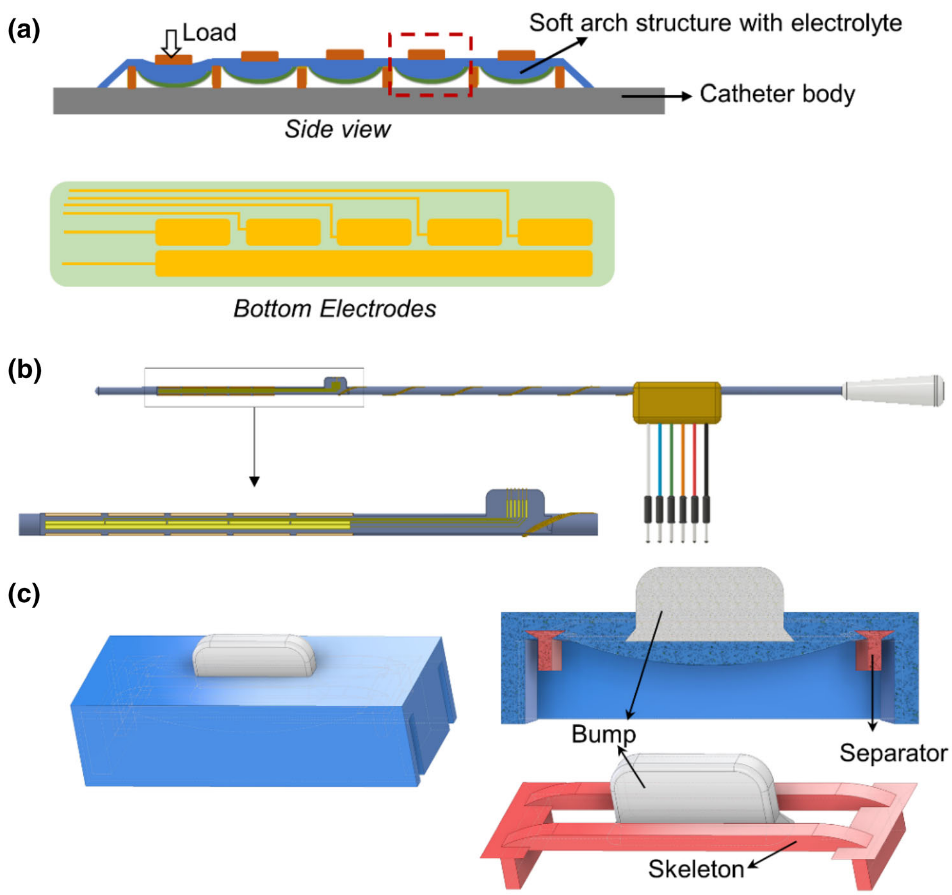

FIGURE 3. (a) Working mechanism of the distributed sensors on the catheter, (b) Schematic of the catheter and bottom electrodes, and (c) Schematic of chamber with electrolyte with half section view and embedded hard structure of the chamber.

based solid electrolyte has excellent mechanical properties as shown in Table 1. It is highly flexible and stretchable. Besides, the ease of making paper substrates of different shapes allows us to easily construct electrolytes of various complicated 3D geometries.

The soft parts with the cured electrolytes were then assembled over the electrodes using sealing glue. Connecting wires (40 AWG, enamel wires) are soldered onto the electrical contact pads on the PI substrate and then wrapped around the catheter. The other ends of the wires are connected to five jumper wires for easy connection to the measurement equipment. The assembled instrumented catheter with five iontronic sensors is shown in Fig. 4f. All the materials used for making the instrumented urethral catheter are specified in Table 2.

Test Platforms

To evaluate the performance of the urethral catheter based on the iontronic sensors, four different test platforms are created and used:

(1) *In vitro load cell experiment setup* The catheter is tested with monotonically increasing forces using a self-designed laboratory test setup as shown in Figs. 5a and 5b. The urethral catheter is fixed at the bottom stage, while varying force is applied from the top through a load cell (ATI, Nano 17). The load cell is fixed on a translational stage, that can move up and down. By moving the load cell down, the force is increased on the sensor. The actual force readings can be recorded by the load cell. The corresponding capacitance response of each sensor is recorded (Rigol DM3068) along with the changing force.

(2) *Cuff test* To create a situation that is somewhat similar to that inside a real urethra, a cuff test is designed as shown in Fig. 5c. An inflatable cuff (AMS 80 Urinary Control System) is wrapped around each sensor. The cuff is generally used to mimic normal sphincter function by opening and closing the urethra during clinical diagnostics. The other end of the cuff is connected to a syringe and a pressure gauge. By moving the

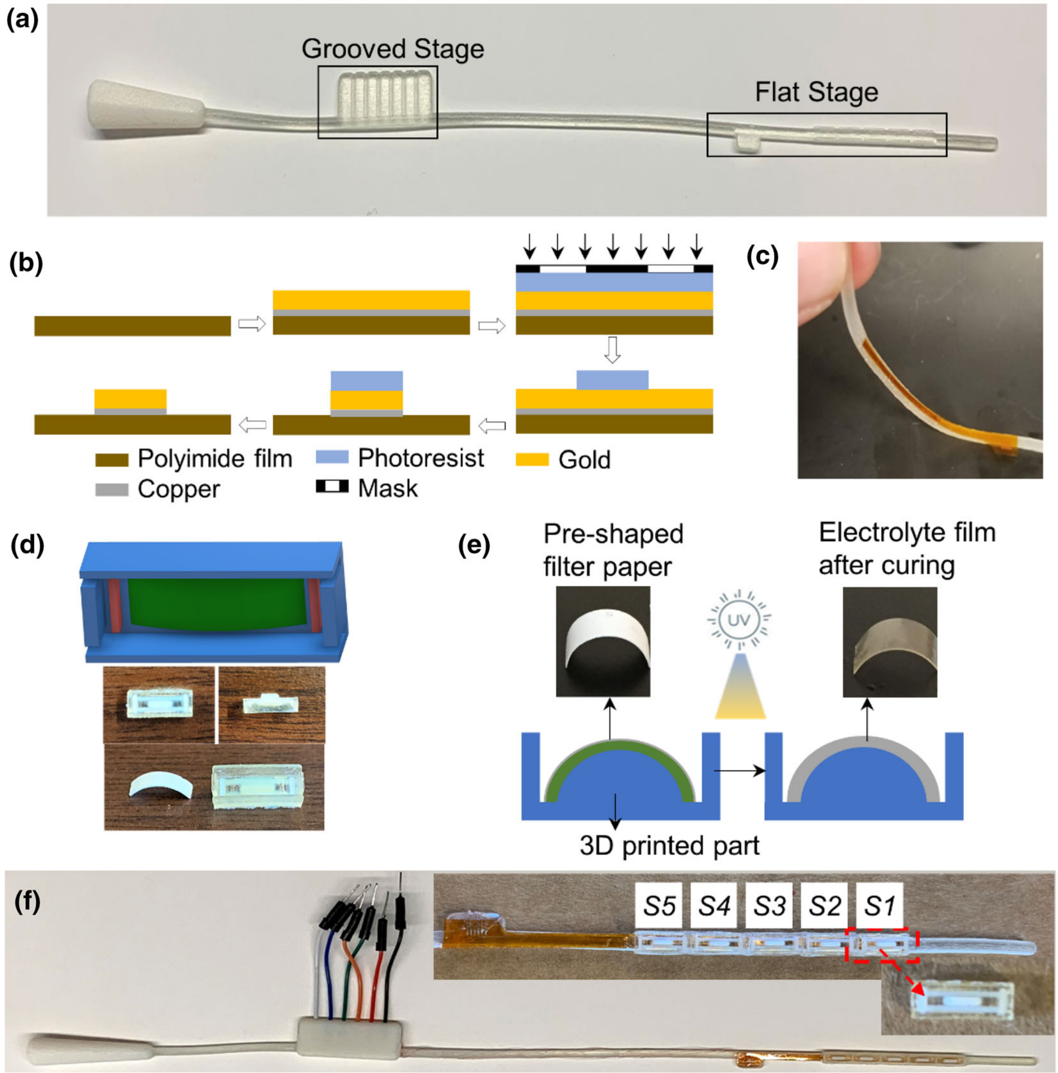

FIGURE 4. (a) 3D printed catheter body, (b) Fabrication process of the bottom electrodes, (c) The electrodes patterned on PI substrate, (d) Schematic and photo of the pre-shaped filter paper and 3D printed part, (e) Fabrication of the electrolyte, and (f) The assembled urethral catheter with 5 iontronic sensors.

TABLE 1. Properties of the paper-based solid electrolyte.

Young's modulus (MPa)	Ultimate tensile strength (MPa)	Toughness (N/m^2)	Configurability
4.6	3.06	20.1×10^6	Various geometries: arch, cylinder, spiral, etc.

piston in the syringe, the cuff can be inflated, and pressure can be increased gradually inside the cuff, thus applying more pressure on the sensor top. The pressure is read from the pressure gauge while the response of the sensors is simultaneously measured.

(3) *In-water test* In order to investigate the influence of parasitic noise on the instrumented catheter, the iontronic sensors were dipped inside water, as shown in Fig. 5d. The capacitance response of each sensor is recorded, both while it is outside water and inside water. The increased capacitance after being put inside water is regarded as the parasitic capacitance.

(4) *Extracted bladder test* The instrumented catheter was tested inside an extracted sheep bladder with

TABLE 2. Materials of the urethral catheter.

Part name	Product name/specification	Manufacturer/provider
Catheter body/electrolyte chamber—soft material	Agilus30/shore hardness 30–35 Scale A	Stratasys
Catheter body/electrolyte chamber—hard part	Vero/ shore hardness 83–86 Scale D	Stratasys
Sealant	Clear Silicone Waterproof Sealant	Loctite
Substrate of the electrodes	Polyimide 75 μm, 300 HPP-ST	Dupont
Connecting wires	40 AWG, enamel wires	Elektrisola
Filter paper	HATF, 0.45 μm	MF-Millipore
Ionic liquid	1-ethyl-3-methylimidazolium tricyanomethanide [EMIM][TCM]	IOLITEC Inc.
Prepolymer	PEG diacrylate (PEGDA, Mw = 575 g/mol) monomers	Sigma-Aldrich
Photo initiator	2-hydroxy-2-methylpropiophenone (HOMPP)	Sigma-Aldrich

the urethra. The catheter is inserted into an extracted bladder as shown in Figs. 5e and 5f, with the five sensors being inside the urethra. The locations of the five sensors in the urethra are marked as S1, S2 ,…, S5 and can be seen in Fig. 5e. An elastic band is wrapped around the urethra from the outside, with one end fixed and the other end connected to a force gauge (Fig. 5f). By gradually moving the force gauge to apply increasing force and stretch the band, pressure can be applied on the sensors. In this test, the stretching force on the elastic band is measured, however, the consequent individual force applied on each of the sensors is not known. The objective of this test is simply to qualitatively evaluate whether the instrumented catheter can provide measurements inside extracted animal tissue.

RESULTS

In Vitro Load Cell Experiment Setup

The response of each sensor to changing forces is shown in Fig. 6a for the *in vitro* load cell tests. Due to the manufacturing imperfections, the calibration curves are slightly different for the five sensors. All the sensor readings increase monotonically with the increasing force applied on the sensors. The sensitivity of the sensors is the in order of 30–50 nF/N, which is more than 1000 times higher than that of conventional capacitive sensors.[3] It should be noted that the sensors have different sensitivities due to manufacturing imperfections and the limited accuracy of the 3D printer that we utilized. Due to the small size of a urethral catheter, more accurate manufacturing is needed in order to get uniform sensitivity. However, we can compensate for the different sensitivities by calibrating all the 5 sensors.

Cuff Test

The capacitance response of the sensor to the increasing pressure inside the cuff is shown in Fig. 6b for the cuff tests. The pressure inside the cuff was increased by moving the piston of the syringe in steps of 10 mmHg up to 120 mmHg. The sensor responses were recorded at each step. All the sensor readings increase monotonically with the increasing pressure inside the cuff. The typical maximum pressure inside the human bladder that needs to be measured is about 100 cm H_2O, which is 74 mmHg, so the device can be fully function up to the maximum expected distributed urethral pressure. It should be noted that the cuff pressure is not the real pressure applied directly on the sensors, due to the cuff not being a circular tube but instead having three portions constituting a triangle. The cross-section of the cuff with the three portions of the triangle is shown in Fig. 5c. The cuff test creates a loading situation similar to that inside the urethra, where the tightening of sphincter muscle of the urethra exerts pressure on the iontronic sensors of the urethra catheter.

In-Water Test

In the case of a traditional parallel plate capacitor, fringe electrical field outside the plates penetrates neighboring materials as shown in Fig. 7a and builds additional capacitance, which is called parasitic capacitance. For *in vivo* applications, due to the large dielectric constant of water and tissues that surround the sensor, the magnitude of the parasitic capacitance can be relatively large (10 s of pico Farads, pF),[2] compared to the small direct response of a traditional capacitive sensor.

Measures could be taken to the remove this parasitic noise. Parasitic capacitance can be compensated by measurement in the absence of loads and subsequent subtraction, but this is only possible when the parasitic noise is static. In biomedical *in vivo* applications of the

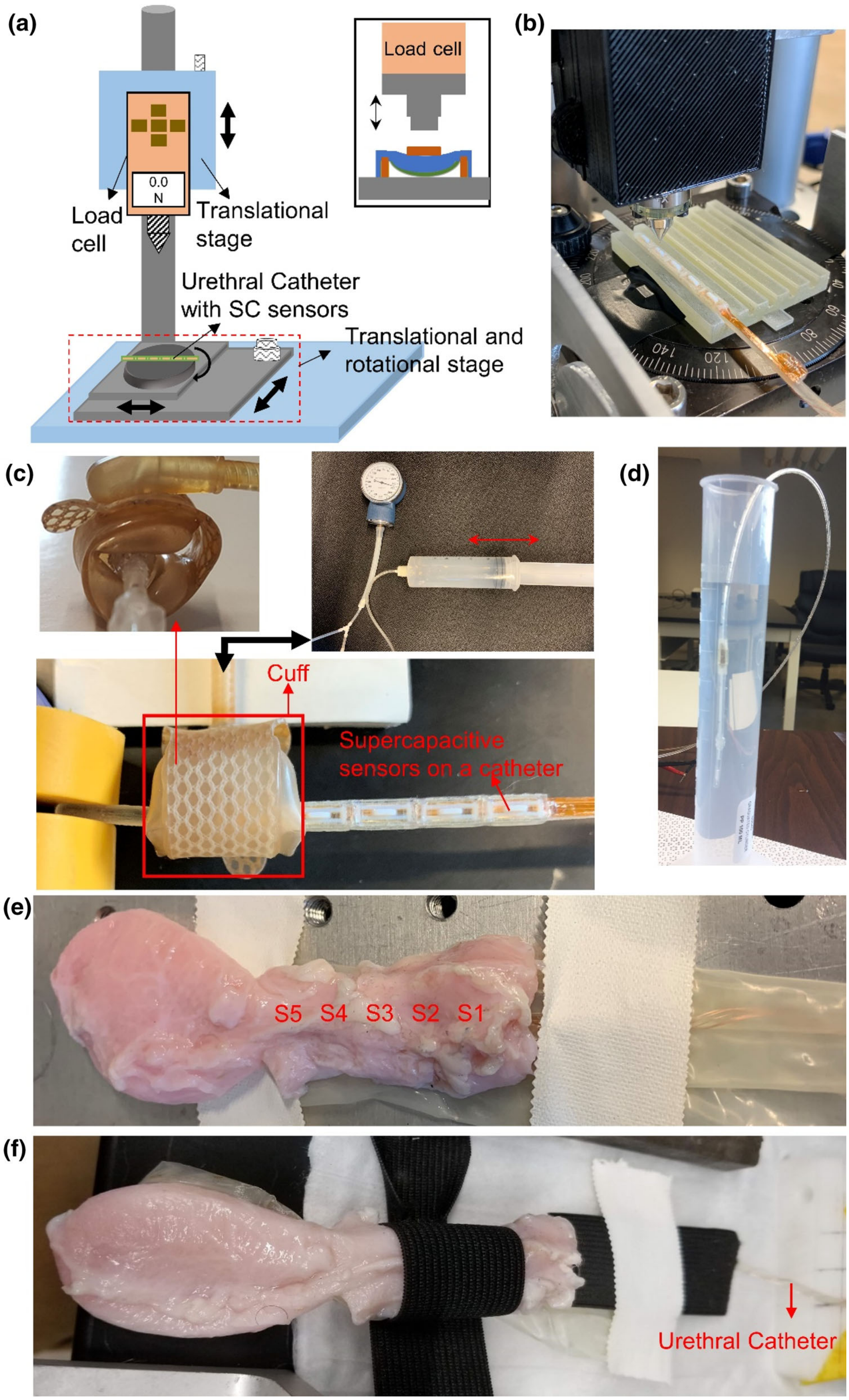

FIGURE 5. Different test platforms for evaluation of the urethral catheter: (a) and (b) the schematic and photo of the *in vitro* load cell experimental setup, (c) Cuff test setup, (d) Inside water test setup, and Urethral catheter inside an extracted bladder with the sensors being inside the urethra (e) before and (f) after loading.

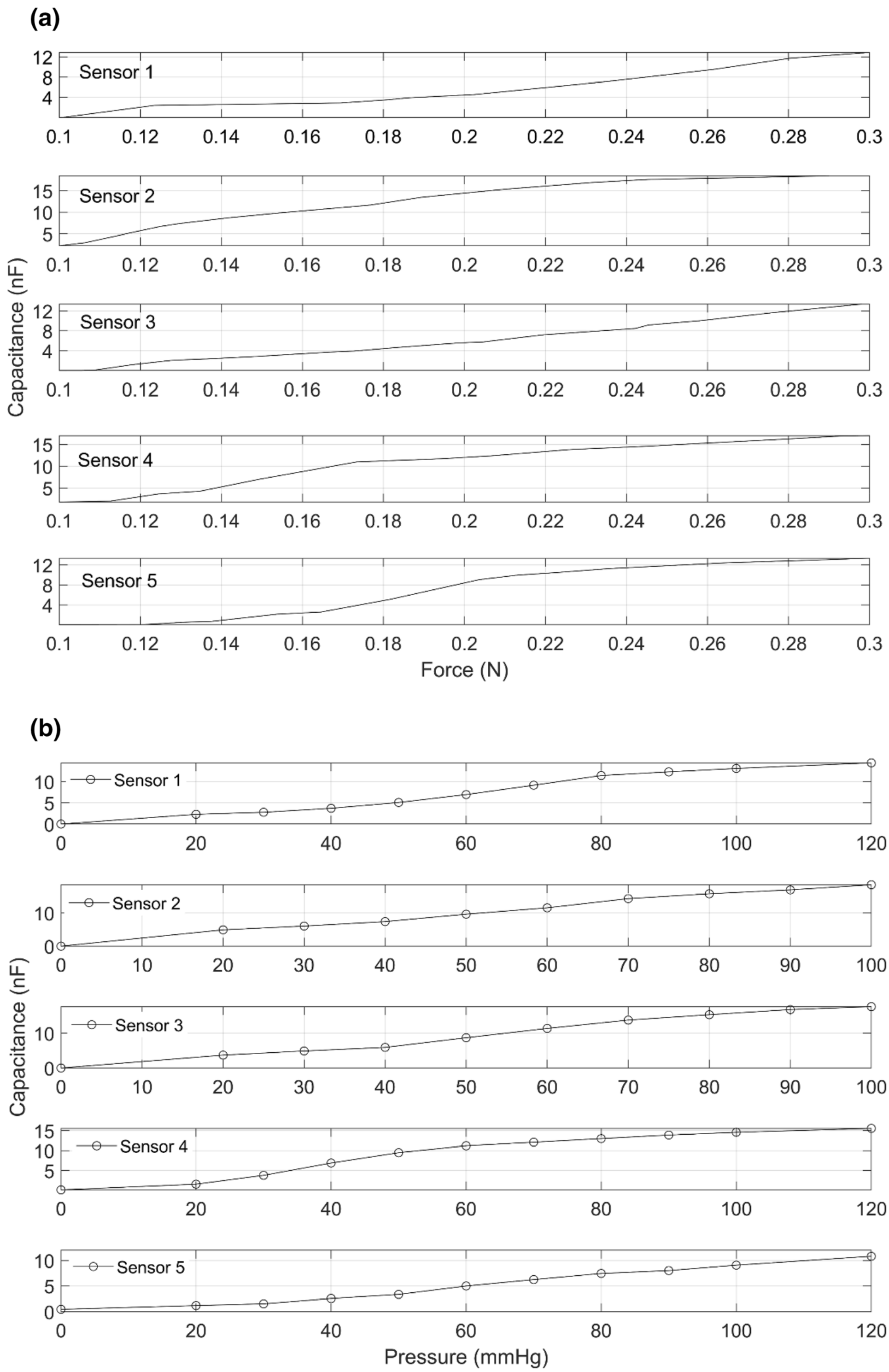

FIGURE 6. Test results: (a) Capacitance responses of the sensors with load cell forces (b) Capacitance responses of the sensors to the pressure cuff tests.

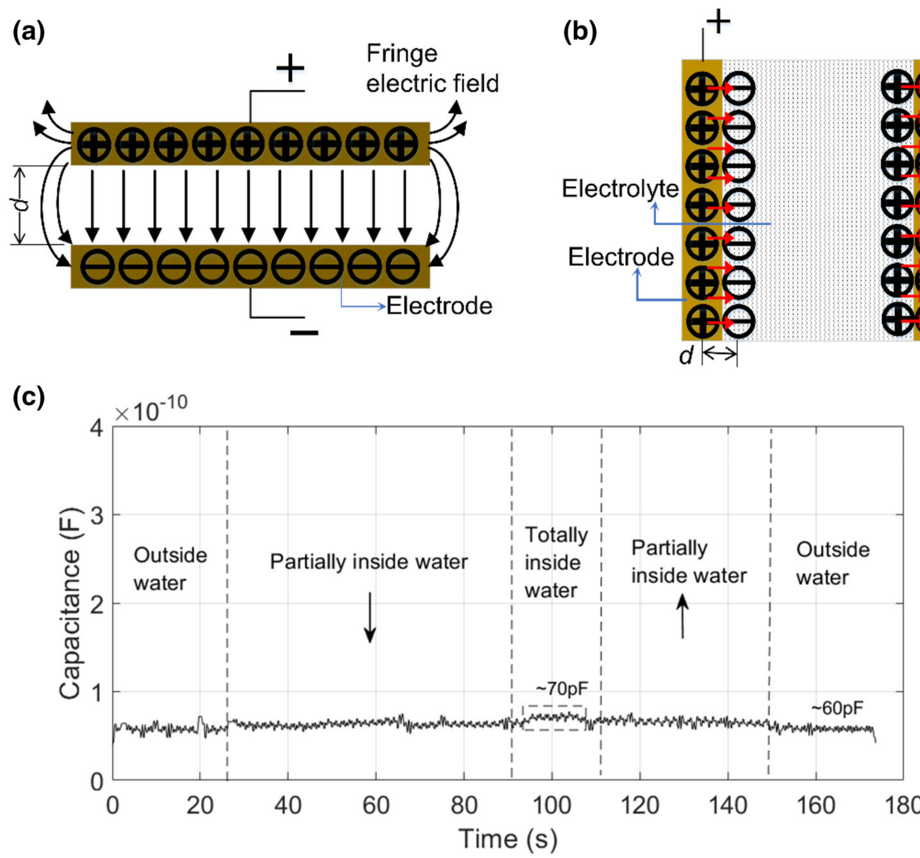

FIGURE 7. Results related to parasitic capacitance: (a) Fringe electric field on a capacitive sensor, (b) Schematic of a iontronic sensor, and (c) Capacitance response outside water and inside water.

capacitive sensors, the interference from human tissues on the fringe electric field of the sensor can cause dynamic parasitic noise, which can greatly pollute the sensor signal and cannot be eliminated by subtraction. Another method has been explored to remove the parasitics by using an active Faraday cage.[2] However, there was still significant parasitic capacitance left and this method involves a complex fabrication process in order to create a highly conductive Faraday cage that surrounds the entire sensor. In a iontronic sensor, the distance between the positive and negative charges at each electrode is of the order of the size of one or two atomic layers (Fig. 7b). Hence, the fringe fields which result in parasitic noise in liquid environment are negligible compared to that on a traditional capacitive sensor. Furthermore, the high capacitance in the supercapacitor overwhelms the parasitic capacitance. Therefore, we hypothesize that when being used in liquid environments like *in vivo* use, the urethral catheter based on iontronic sensing does not experience additional parasitic noise.

The parasitic capacitance test result is shown in Fig. 7c. After dipping into water, the capacitance increases by 10 pF, which could be caused by the water pressure after dipping into water. That means the parasitic noise is negligible compared to the high sensitivity of each sensor (30–50 nF/N). This verifies that the catheter is immune from parasitic noise induced by human tissue for *in vivo* applications.

Extracted Bladder-Urethra Test

Figure 8 shows the responses of the sensors during an *ex vivo* test involving an extracted sheep bladder and urethra. As the stretching force applied on the elastic band increases, the tightening of the band exerts pressure on the tissue. The responses of all the five sensors increase monotonically. The shape of the external tissue of the urethra is not uniform, which creates a non-uniform pressure distribution on the instrumented urethra catheter from the tightening elastic band as shown in Fig. 8a. The force was applied on the urethral tissue by wrapping a wide cloth band around it and pulling on one side of the band through a force gauge. Even when no force is applied for stretching the elastic band, the wrapping of the elastic band around the urethra alone can also lead to a capacitance change because it does apply some force on the tissue. That's why each sensor has a small dif-

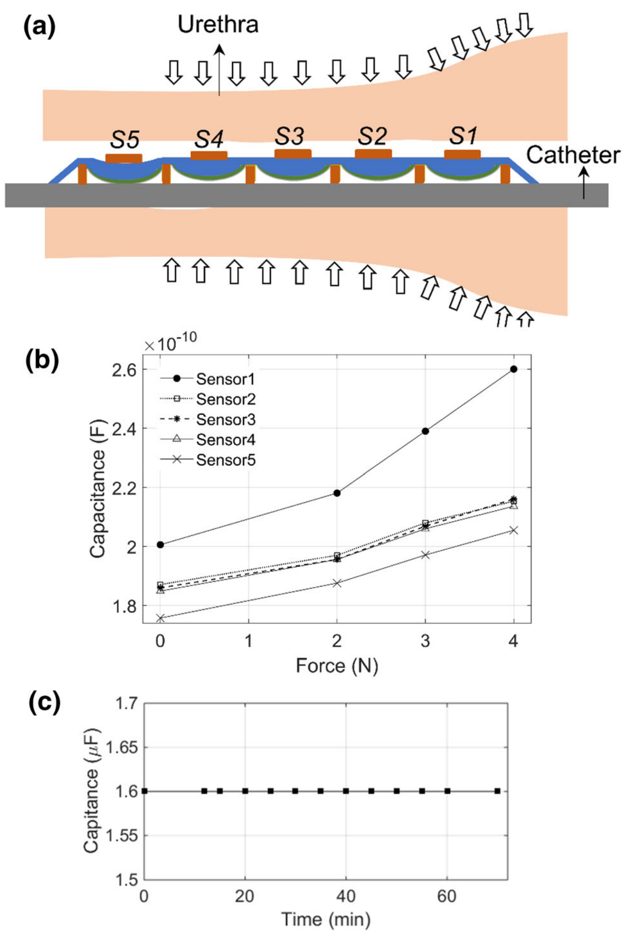

FIGURE 8. (a) The instrumented urethra catheter inserted into the extracted sheep urethra, (b) Sensor response inside the urethra (c) Drifting performance test result.

ferent amount of initial capacitance. When the band is pulled with a force, it squeezes the tissue inside the band. Since the tissue is not uniformly sized across the width of the band (see Fig. 8a), the portions of the tissue that are larger experience more force. Since sensors 2, 3 and 4 are located in the middle of the urethra (which is a thinner portion of the urethra), while sensor 1 is located under the thicker tissue part which experiences more force, the response of sensor 1 is significantly higher than that of sensors 2–4. Sensor 5 is outside the wrapping zone of the elastic band, so the force is not directly applied on top of sensor 5. However, sensor 5 is also covered by the urethra tissue, even if the band is not directly above it. When force is applied to stretch the elastic band, the tissue around sensor 5 also experiences a little bit of pressure due to the tissue getting squeezed from the force applied at the other portions of it.

A test of the drifting performance of the iontronic sensors was conducted on a single static iontronic cell of similar design. The test result is shown in Fig. 8c. The capacitance readings were recorded every few minutes for a total time of 70 min to check if there are any continuous slow drifts in the static reading. The capacitance change is negligible for the duration of 70 min of recording, as seen in Fig. 8c.

DISCUSSION

A multi-sensor flexible instrumented catheter for measurement of distributed pressure inside the urethra was developed. The catheter had modular force sensors that were distributed on a flattened portion of the catheter. The developed instrumented catheter has important clinical applications in urodynamic testing and potentially in other *in vivo* biomedical catheter applications. Iontronic force sensors were designed and fabricated on the catheter using a combination of MEMS technologies and 3D printing. Experimental *in vitro* evaluation was conducted using a self-designed setup with a load cell and then later inside an inflatable cuff. The influence of parasitic noise was investigated by dipping the sensor inside water and measuring its resulting capacitance change. Also, *ex vivo* tests were conducted inside an extracted bladder and urethra.

The developed urethral catheter is unique. No other catheter in literature successfully incorporates multiple force sensors for measurement of distributed pressure. Also, the use of iontronic sensing and fabrication of sensors integrated with the catheter using 3D printing are both innovative and practically valuable contributions. It was important to determine whether the developed catheter could meet performance requirements such as adequate force sensitivity, ability to measure forces inside animal tissue such as extracted bladder and urethra, and ability to measure in liquid environments without suffering from parasitic capacitance.

A number of experiments were carried out to evaluate the performance of the catheter, as described in the section on Materials and Methods. First, the sensors on the catheter showed excellent sensitivity when evaluated in a load cell test rig, using an expensive multi-axes reference load cell. The sensors were found to have a sensitivity of 30–50 nF/N which is more than 1000 times larger than traditional capacitive sensors on similar-sized catheters.[2–4] Further, the sensors on the integrated catheter showed almost no change in capacitance when dipped fully in water. This is a very significant advantage compared to other urethral catheters in literature.[3,4] Finally, the sensors were also able to measure forces well when used inside an extracted urethra from a sheep. The sensors showed almost no change in capacitance from being inserted into the extracted urethra but clearly showed sensitive responses to application of force on the urethra. All of these results together indicate that the developed instrumented catheter is a viable device for actual use and could be taken up for further evaluations in animal tests followed by human subject tests.

While other researchers have previously explored iontronic sensors from a fundamental research perspective,[19–21] this is the first time that such iontronic sensors have been integrated into an instrumented catheter and utilized in a medical device for a real-world Urology application. Previous instrumented catheters found in the real world include the Medtronic Manoscan System.[17] The Manoscan catheter is designed for the esophagus, and not for Urology applications. Further, it is a very expensive system with a cost well above $50,000. Only a few real-world applications[37] have previously utilized the specific benefits of iontronic sensors, namely their ultra-high sensitivity and their immunity to parasitic capacitance in liquids in a critical fashion. When compared with other catheters that have been utilized in Urology, the previous catheters utilized in human subject tests have all involved balloon-based devices and have only utilized a single balloon sensor at the tip of the catheter. The disadvantages of using balloon sensors include the need for significant external pneumatic controls, low bandwidth and the inability to sufficiently miniaturize for incorporating multiple sensors on one device.

The success of the preliminary experiments conducted in this paper opens up the possibility that future work could be conducted that involves *in vivo* evaluation of the catheter in an IACUC approved live sheep study and later in an IRB approved clinical human study.

ACKNOWLEDGMENTS

This work was supported in part by Mn-Reach, a NIH research Evaluation and Commercialization Hub, and by the National Science Foundation through Grant EFRI 1830958.

CONFLICT OF INTEREST

A portion of the work reported in this paper has been protected through a patent filing. The pending patent will belong to the University of Minnesota which has a standard royalty sharing agreement with university employees, in case any royalties are earned from the licensing of said patent.

REFERENCES

1. Abrams, P., J. G. Blaivas, S. L. Stanton, and J. T. Anderson. The standardization of terminology of lower urinary tract function recommended by the international continence society. *Int. Urogynecol. J.* 1:45–58, 1990.
2. Ahmadi, M., R. Rajamani, and S. Sezen. Transparent flexible active faraday cage enables in vivo capacitance measurement in assembled microsensor. *IEEE Sens. Lett.* 1:289–313, 2017.
3. Ahmadi, M., R. Rajamani, G. Timm, and A. S. Sezen. Flexible distributed pressure sensing strip for a urethral catheter. *J. Microelectromech. Syst.* 24:1840–1847, 2015.
4. Ahmadi, M., R. Rajamani, G. Timm, and S. Sezen. Instrumented urethral catheter and its ex vivo validation in a sheep urethra. *Meas. Sci. Technol.* 28:1–10, 2017.
5. Conway, B. E. Electrochemical Supercapacitors. New York: Springer, p. 714, 1999.
6. Dario, P., D. Femi, and F. Vivaldi. Fiber-optic catheter-tip sensor based on the photoelastic effect. *Sens. Actuator* 12:5–47, 1987.
7. Diokno, A. C., B. M. Brock, M. B. Brown, and A. R. Herzog. Prevalence of urinary incontinence and other urological symptoms in the noninstitutionalized elderly. *J. Urol.* 136:1021–1025, 1986.
8. Esashi, M., H. Komatsu, T. Matsuo, M. Takahashi, T. Takishima, K. Imabayashi, and H. Ozawa. Fabrication of catheter-tip and sidewall miniature pressure sensors. *IEEE Trans. Electron. Dev.* 29:57–63, 1982.
9. Feneley, R. C. L., I. B. Hopley, and P. N. T. Wells. Urinary catheters: history, current status, adverse events and research agenda. *J. Med. Eng. Technol.* 39:459–470, 2015.
10. Feneley, R. C. L., A. M. Sheperd, P. H. Powell, and J. Blannin. Urinary incontinence: prevalence and needs. *Br. J. Urol.* 51:493–496, 1979.
11. French, P. J., D. Tanase, and J. F. L. Goosen. Sensors for catheter applications. In: Sensors Applications. 2008, pp. 339–380.
12. Hannestad, Y. S., G. Rortveit, H. Sandvik, and S. Hunskaar. A community-based epidemiological survey of female urinary incontinence: the Norwegian EPINCONT study. Epidemiology of Incontinence in the County of Nord-Trøndelag. *J. Clin. Epidemiol.* 53:1150–1157, 2000.
13. Herzog, A. R., and N. H. Fultz. Prevalence and incidence of urinary incontinence in community-dwelling populations. *J. Am. Geriatr. Soc.* 38:273–281, 1990.
14. Jin, M. L., S. Park, Y. Lee, J. H. Lee, J. Chung, J. S. Kim, J. S. Kim, S. Y. Kim, E. Jee, D. W. Kim, J. W. Chung, S. G. Lee, D. Choi, H. T. Jung, and D. H. Kim. An ultrasensitive, visco-poroelastic artificial mechanotransducer skin inspired by Piezo2 protein in mammalian merkel cells. *Adv. Mater.* 29:1–9, 2017.
15. Li, R., Y. Si, Z. Zhu, Y. Guo, Y. Zhang, N. Pan, G. Sun, and T. Pan. Supercapacitive iontronic nanofabric sensing. *Adv. Mater.* 29:1–8, 2017.
16. Malik, P. Grossman's cardiac catheterization, angiography, and intervention. *Can. J. Cardiol.* 23:602, 2007.
17. Medtronic. Manoscan ESO high resolution manometry system. https://www.medtronic.com/covidien/en-us/products/motility-testing/manoscan-eso-high-resolution-manometry-system.html#manoscan-eso-high-resolution-manometry-catheters.
18. Meena, K. V., R. Mathew, and A. R. Sankar. Design and optimization of a three-terminal piezoresistive pressure

sensor for catheter based in vivo biomedical applications. *Biomed. Phys. Eng. Express* 3:045003, 2017.

[19]Nawi, M. N. M., A. A. Manaf, M. F. A. Rahman, M. R. Arshad, and O. Sidek. One-side-electrode-type fluidic-based capacitive pressure sensor. *IEEE Sens. J.* 15:1738–1746, 2015.

[20]Nie, B., R. Li, J. Cao, J. D. Brandt, and T. Pan. Flexible transparent iontronic film for interfacial capacitive pressure sensing. *Adv. Mater.* 27:6055–6062, 2015.

[21]Nie, B., S. Xing, J. Brant, and T. Pan. Droplet-based interfacial capacitive sensing. *Lab Chip* 12:1110–1118, 2012.

[22]Nygaard, I., T. Girts, N. H. Fultz, K. Kinchen, G. Pohl, and B. Sternfeld. Is urinary incontinence a barrier to exercise in women? *Obstet. Gynecol.* 106:304–314, 2005.

[23]Otmani, R., N. Benmoussa, and B. Benyoucef. The thermal drift characteristics of piezoresistive pressure sensor. *Phys. Procedia* 21:47–52, 2011.

[24]Rafii-Tari, H., C. J. Payne, and G.-Z. Yang. Current and emerging robot-assisted endovascular catheterization technologies: a review. *Ann. Biomed. Eng.* 42:697–715, 2014.

[25]Resplande, J., S. Gholami, H. Bruschini, and M. Srougi. Urodynamic changes induced by the intravaginal electrode during pelvic floor electrical stimulation. *Neurourol. Urodyn.* 22:24–28, 2003.

[26]Schafer, W., P. Abrams, L. Liao, A. Mattiasson, F. Pesce, A. Spangberg, A. Sterling, N. R. Zinner, and P. Kerrebroeck. Good urodynamic practices: uroflowmetry, filling cystometry, and pressure-flow studies. *Neurourol. Urodyn.* 21:261–274, 2002.

[27]Sharma, T., K. Aroom, S. Naik, B. Gill, and J. X. J. Zhang. Flexible Thin-Film PVDF-TrFE Based Pressure Sensor for Smart Catheter Applications. *Ann. Biomed. Eng.* 41:744–751, 2013.

[28]Taufik, M., and P. Jain. Role of build orientation in layered manufacturing: a review. *Int. J. Manuf. Technol. Manag.* 27:47–73, 2014.

[29]Thom, D. H., M. N. Haan, and S. K. Van Den Eeden. Medically recognized urinary incontinence and risks of hospitalization, nursing home admission and mortality. *Age Ageing* 26:367–374, 1997.

[30]Van Oyen, H., and P. Van Oyen. Urinary incontinence in belgium: prevalence, correlates and psychosocial consequences. *Acta Clin. Belg.* 57:207–218, 2002.

[31]Wagner, T. H., and T.-W. Hu. Economic costs of urinary incontinence in 1995. *Urology* 51:355–361, 1998.

[32]Webb, R. J., P. D. Ramsden, and D. E. Neal. Ambulatory monitoring and electronic measurement of urinary leakage in the diagnosis of detrusor instability and incontinence. *Obstet. Gynecol. Surv.* 47:148–152, 1992.

[33]Wu, X., M. Hu, J. Shen, and Q. Ma. A miniature piezoresistive catheter pressure sensor. *Sens. Actuat: A* 35:197–201, 1993.

[34]Yarnell, J. W. G., and A. S. StLeger. The prevalence, severity and factors associated with urinary incontinence in a random sample of the elderly. *Age Ageing* 8:81–85, 1979.

[35]Zhang, Y., R. Rajamani, and S. Sezen. Novel supercapacitor-based force sensor insensitive to parasitic noise. *IEEE Sens. Lett.* 1:2–5, 2017.

[36]Zhang, Y., S. Sezen, M. Ahmadi, X. Cheng, and R. Rajamani. Paper-based supercapacitive mechanical sensors. *Sci. Rep.* 8:16284, 2018.

[37]Zhang, Z., Z. Zhu, B. Bazor, S. Lee, Z. Ding, and T. Pan. FeetBeat: a flexible iontronic sensing wearable detects pedal pulses and muscular activities. *IEEE Trans. Biomed. Eng.* 66:3072–3079, 2019.

[38]Zhu, Z., R. Li, and T. Pan. Imperceptible epidermal-iontronic interface for wearable sensing. *Adv. Mater.* 30:1–9, 2018.