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Communication

Micropatterned elastic ionic polyacrylamide hydrogel for low-voltage capacitive and organic thin-film transistor pressure sensors

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ABSTRACT

Electronic skins (E-skins) have attracted great research interest because of their promising applications in stretchable optoelectronics, soft robotics, and personalized healthcare devices. However, it remains a great challenge to fabricate E-skin devices that meet strict practical requirements such as high sensitivity, low-power operation and noise-proof ability. Here, we developed a novel elastic ionic polyacrylamide hydrogel (EIPH) with a high capacitance for the development of low-voltage organic thin-film transistor (OTFT) pressure sensors. The EIPH was prepared by photopolymerization of an acrylamide monomer in an aqueous solution of poly (acrylic acid) and $CaCl_2$ and was then *in situ* micropatterned on an indium-tin oxide electrode. The fabricated capacitive sensor with 10-μm-wide EIPH micropillar structures achieved a high sensitivity of 2.33 kPa^{-1} with a capacitance sensitivity of 103.8 nF/kPa. This capacitance sensitivity is more than 100 times higher than that of conventional capacitive pressure sensors due to the formation of an electrical double layer. The micropatterned EIPH was adopted as a dielectric layer in the fabrication of the OTFT-based pressure sensors. Such an EIPH-based OTFT pressure sensor not only greatly enhanced the sensitivity, *i.e.,* 7.7 times higher than its capacitive counterparts, but also largely reduced the operation voltage to 2 V.

1. Introduction

Human skin enables us to perceive the changes in the contact pressure and the shapes and textures of objects. To mimic these abilities, various flexible pressure sensors have been developed [1–5], especially for the applications of electronic skin (E-skin) [6,7], soft robotics [8,9], flexible touch displays [10–13], energy harvesting [14,15], and healthcare [16–20]. The working mechanisms of these pressure sensors include piezoelectricity [20–26], capacitance [11,16,17,27,28], triboelectricity [10,29,30], and piezoresistivity [31–34], in which an applied force is converted into electrical signals that can be directly read. Particularly, capacitive sensors have recently gained enormous success in consumer electronics because of their distinct advantages of high electrical sensitivity, low-power consumption, compact circuit layout, and a simple device fabrication process [35,36]. Substantial advancements have been achieved in recent years, aiming at improving the sensitivity of capacitive pressure sensors to detect ultralow pressure variations [2,17]. For example, Bao's group proposed to pattern polydimethylsiloxane (PDMS) into microstructures to enhance the sensitivity of PDMS-based pressure sensors [27,28]. Kwon et al. demonstrated a highly sensitive piezocapacitive pressure sensor using a three-dimensional (3D) microporous Ecoflex dielectric layer [17]. However, the sensitivities of those pressure sensors are lower than 1 kPa^{-1}, even in the low-pressure range.

One promising solution to this issue is to further integrate the microengineered elastomer into organic thin-film transistors (OTFTs) for signal amplification and sensing-mechanism diversification [1–3,37]. Indeed, given by their excellent flexibility, light-weight, and potentially low-cost [38], OTFT devices have been broadly demonstrated for various flexible sensing applications [39–41]. For instance, an OTFT-based pressure sensor was first proposed by Someya et al. [42], in which an OTFT was employed as an electronic readout element for conductive rubber pressure sensing components. The sensitivity of the sensor, however, is not high due to the limitation of the integrated conductive rubber sensing element. Later, Bao et al. incorporated microstructured PDMS as a dielectric layer into a flexible OTFT-based pressure sensor [5] and achieved a high sensitivity up to 8.2 kPa^{-1} with the ability of subtle pressure detection [18]. More recently, the sensitivity of the OTFT-based pressure sensor was enhanced to an ultrahigh level of 192 kPa^{-1} with well tunability by using a novel suspended gate OTFT

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structure [19]. Nevertheless, the operation voltage of most OTFT-based pressure sensors is usually high and ranges from tens to even hundreds of volts [5,18,19], which limits their practical application particularly in portable and wearable electronic devices.

Enhancing the capacitance of the dielectric layer is a well-proved way to reduce the operation voltage of OTFT devices [43]. Notably, the ionic materials, consisting of a substantial number of mobile cations and anions, can form an electrical double layer (EDL) with exceptionally high unit area capacitance, once sandwiched between two electrodes [35,44–46]. The capacitances of the ionic materials are typically more than 1000 times greater than that of traditional parallel-plate devices [35,36,47–50]. Recently, Chang's group utilized an ionic liquid as the electrode to form an EDL structure and then adopted PDMS as an elastomer to fabricate an EDL-based capacitive pressure sensor [49,50]. The sensor revealed a high-pressure sensitivity of 9.55 kPa^{-1}. However, the high sensitivity was only maintained in very narrow pressure range, *i.e.,* from 0 to 0.2 kPa, and its capacitance was not high because of the utilization of PDMS for the dielectric layer. Pan's group demonstrated an EDL-based capacitive pressure sensor by adhering an iontronic film on a flexible electrode and then suspending it upon another flexible electrode [36,47]. Such a flexible sensor attained a sensitivity of 3.1 nF/kPa. They also reported an EDL-based iontronic microdroplet array device for capacitive pressure sensing, whose sensitivity is relatively lower, *i.e.* 0.43 nF/kPa [51]. For those dielectric materials, the nonlinear response and complex structure limit their application for further development of OTFT devices. By mimicking Piezo2 protein in Merkel cells, Kim's group patterned ionic elastomer (ionic-liquid-loaded ionic thermoplastic polyurethane) into pillar-shapes and fabricated a high-performance capacitive sensor covering a wide range of pressures [52,53]. Recently, we demonstrated an EDL-type suspended gate OTFT pressure sensor, in which a polyelectrolyte composite was used to prepare the dielectric layer of the flexible OTFT [48]. The operation voltage of this OTFT sensor is 5 V. However, the fabrication of this type of sensor with dual-layer dielectric layer is sophisticated, and there is still space to further lower the operation voltage because of the low dielectric constant of air.

Inspired by the studies of Suo's group who developed an elastic polyacrylamide (PAAm) composite hydrogel [54], a large number of studies have adopted elastic hydrogels for tactile sensing devices fabrication [12,13,55–59]. However, no report exists regarding the utilization of an elastic hydrogel as a dielectric layer for capacitive or OTFT-based pressure sensing device development, which may have the potential to enhance the performance of hydrogel-based capacitive pressure sensors. Here, we present a photocrosslinkable elastic ionic polyacrylamide hydrogel (EIPH) to fabricate high-sensitivity low-power-consumption capacitive and OTFT pressure sensors. An in-house digital ultraviolet (UV) lithography technology is applied to *in situ* engineer the EIPH, by precisely controlling the photopolymerization and optically patterning the hydrogel in micrometer scale, on an electrode for direct printing of capacitive pressure sensors. Such capacitive pressure sensors show extraordinarily high sensitivity, long-time stability and a good noise-proof property. Moreover, the micropatterned EIPH can be adopted in an OTFT as a dielectric layer for the fabrication of pressure sensors. Experimental results show that such EIPH-based OTFT sensors not only have extraordinarily high sensitivity but also exhibit a low operation voltage.

2. Material and methods

2.1. Materials

Acrylamide, N,N-methylenebisacrylamide (MBA), 2-hydroxy-4′-(2-hydroxyethoxy)-2-methylpropiophenone (Irgacure 2959), and poly (acrylic acid) (PAA, $Mw = 100\ 000\ g\ mol^{-1}$, 35 wt% aqueous solution) were purchased from Sigma-Aldrich. Poly[4,8-bis(5-(2-ethylhexyl) thiophen-2-yl)benzo[1,2-b;4,5-b']dithiophene-2,6-diyl-*alt*-(4-(2-ethylhexyl)-3-fluorothieno[3,4-b]thiophene-)-2-carboxylate-2–6-diyl)] (PTB7-Th) was obtained from 1-Material Inc. Poly(indacenodithiophene-*co*-benzothiadiazole) (PIDT-BT) and tetracyanoquinodimethane (TCNQ) were purchased from Derthon Optoelectronic Materials Science Technology Co., Ltd. and J&K Scientific Ltd. (Beijing), respectively. All the materials were used as received. Other reagents and raw materials were commercially obtained by Sigma-Aldrich Inc., or Adamas-beta Ltd., and were used without further purification. Deionized (DI) water with a resistance of 18 MΩ cm was used in all experiments.

2.2. Micropatternable PAAm solution preparation

Acrylamide monomer (1.8 g) was dissolved in 2.3 g of DI water. Then, 0.26 g of PAA, 0.1 g of Irgacure 2959 (photoinitiator), 0.0072 g of MBA (crosslinker), and 0.0125 g of $CaCl_2$ were added stepwise until fully dissolved under stirring. The prepared photoresist solution was stored for use.

2.3. Capacitive pressure sensor fabrication

The ITO/polyethylene terephthalate (PET) electrode with a thickness of 125 μm was cleaned in an ultrasonic cleaner by immersion into the following solvents: acetone, water, and isopropyl alcohol, each for 30 min. The electrode was then blow dried with a nitrogen gun. The photoresist solution was dropped on the ITO/PET electrode and micropatterned with an in-house optical maskless exposure setup *via* UV light (365 nm). The intensity of the UV light was 15.65 mW/cm^2, and the exposure time was 40 s. After the patterning process, the micropatterned EIPH (area: 1 cm^2) was laminated with another ITO/PET electrode to prepare the capacitive pressure sensors. Then, the sensors were photoannealed by an 8 W UV light for 30 min to reach the desired elasticity.

2.4. OTFT-based pressure sensor fabrication

All the OTFT pressure sensors were prepared on 175-μm-thick PET. After cleaning this plastic substrate, 50-nm-thick gold (Au) as the source/drain (S/D) electrodes was deposited on top of PET by thermal evaporation *via* a shadow mask with a channel width and length of 22.5 mm and 0.15 mm ($W/L = 150$), respectively. This evaporation process was performed under a high vacuum ($\sim 2 \times 10^{-4}$ Pa), and the evaporation rate for Au was approximately 0.5 Å/s. Prior to the semiconductor deposition, the Au S/D electrodes on the PET substrates were treated with UV-O_3 for 15 min and then transferred into a nitrogen-filled glovebox. Thereafter, the organic semiconductor layer was fabricated by spin-casting (2000 rpm, 60 s) a PTB7-Th solution (dissolved in chlorobenzene with a concentration of 3 mg/mL) or a blend PIDT-BT: TCNQ solution (PIDT-BT: TCNQ = 49:1 by weight dissolved in chlorobenzene with a total concentration of 5 mg/mL) on the PET/Au surface, followed by annealing at 80 °C for 20 min. Subsequently, a thin PAA protective layer was prepared on the semiconductor layer. Then, the device was again thermally treated at 80 °C for 20 min. The micropatterned PAAm composite hydrogel was also prepared on the ITO-coated PET following the same procedure described above. Finally, this flexible gate/dielectric part was laminated under ambient conditions by pressing the electrodes against each other and fixing by tape in the device.

2.5. Characterizations and device tests

The microstructures of PAAm composite hydrogels were measured by 3D laser scanning confocal microscope (VK-X200, KEYENCE, Japan) and scanning electron microscopy (JEOL Model JSM-6490). The former is a non-contact laser scanning imaging machine. The magnification of the objective lens used for scanning was 50×.

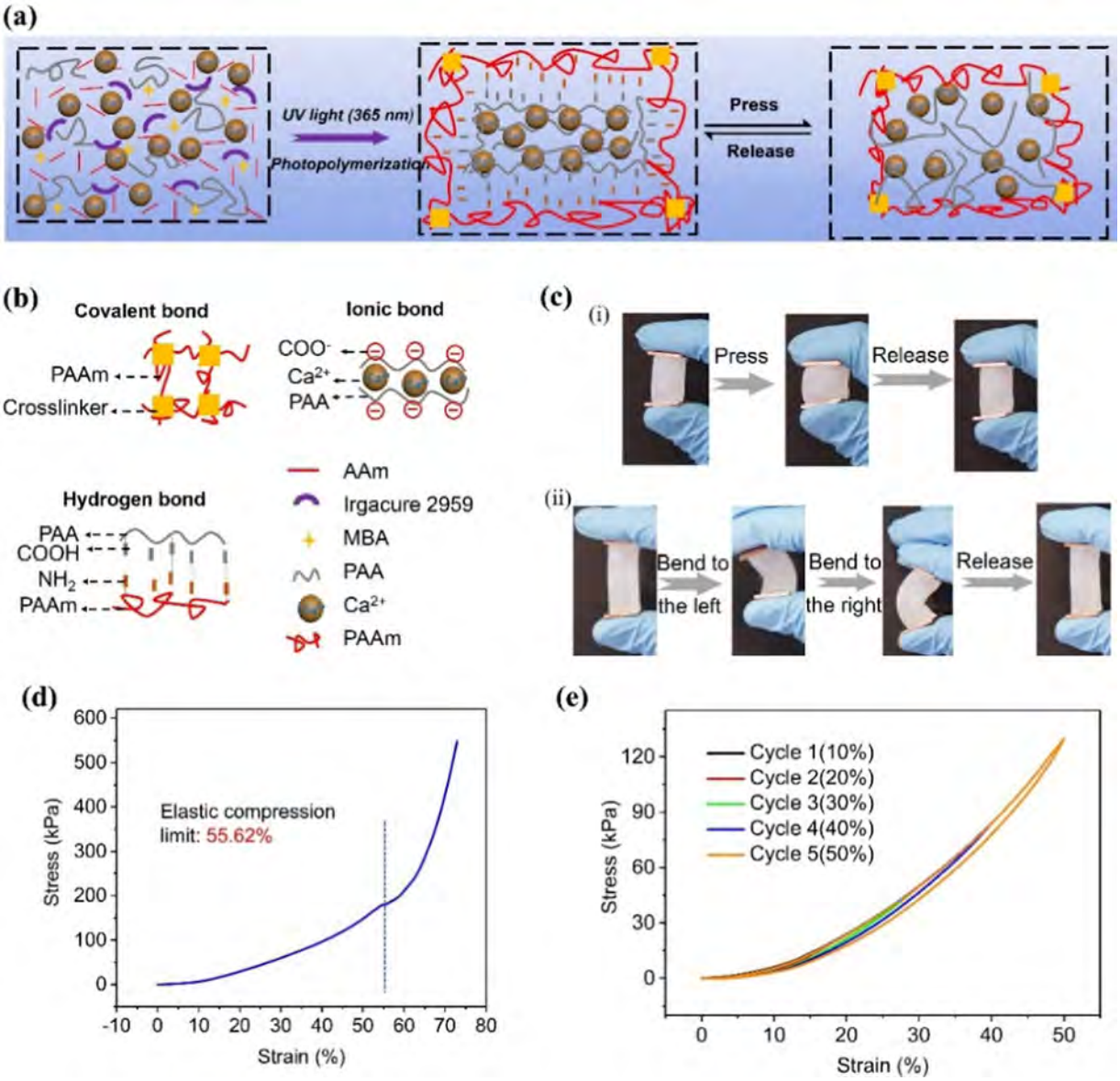

Fig. 1. Photopolymerization and elastic property of the EIPH. (a) Schematic of the photopolymerization process and the reversible breaking/forming of secondary bonds with/without pressure. (b) Three kinds of bonds in the EIPH: covalent bonds between PAAm, ionic bonds between PAA and Ca^{2+}, and hydrogen bonds between PAAm and PAA. (c) Demonstration of the highly elastic property of the EIPH under pressing (top panel) and bending (bottom panel) conditions. (d) The elastic compression limit of EIPH. (e) The measured stress–strain curves of the EIPH with the cycles of the reversible strain changes from 0% to 10%, 20%, 30%, 40%, and 50%, respectively.

Fourier-transform infrared spectroscopy (FTIR) spectra were recorded by a Bruker Vector Spectrometer (Tensor II). The compressive stress–strain measurements were performed using a tensile-compressive tester (Instron-5942 with a 500 N sensor) in air. The rate of compression was kept constant as 2% s^{-1} with respect to the original height of the hydrogel, roughly 5 mm min^{-1}.

The capacitances were measured by LCR-819 (GW Instek, Taiwan, 12 Hz-100 kHz, with accuracy of 0.5%) with an applied voltage of 0.1 V. The electrical characteristics of all the OTFT pressure sensors were measured in ambient air by using an Agilent 4155 C semiconductor parameter analyser. The force applied on the sensors was precisely controlled with a force gauge (Model No.: JSV-H1000).

3. Results and discussion

Fig. 1a shows the scheme of the photopolymerization process to prepare the EIPH with UV light. Specifically, monomers of acrylamide are photopolymerized with a photoinitiator (Irgacure 2959) and crosslinked by the crosslinker (N,N-methylenebisacrylamide, MBA) *via* radical polymerization [59,60], which leads to the formation of 3D networks through covalent bonds. Meanwhile, the PAAm chains will interact with poly(acrylic acid) (PAA) *via* hydrogen bonds, thereby forming a double-network hydrogel. Moreover, the PAA chains are simultaneously crosslinked by Ca^{2+} *via* electrostatic forces (*i.e.,* ionic bonds), which can further enhance the mechanical strength of the composite hydrogel [59]. Such an EIPH with the three kinds of bonding forces is shown in Fig. 1b. Notably, the ionic bonds and hydrogen bonds in the EIPH can dynamically break/form during the process of applying/releasing pressure along with energy dissipation [54,58,59]. Particularly, the covalent crosslinked PAAm networks play an essential role in maintaining the initial geometry of the 3D hydrogel structure, while the ionic bonds and hydrogen bonds render the superelasticity of the EIPH. Fig. S1 gives the FTIR spectra of EIPH, PAA and PAAm hydrogel. The peak located at 1702 cm^{-1} is the typical carbonyl group of

PAA [61]. For PAAm hydrogel, the peak at 1655 cm^{-1} and the shoulder at 1605 cm^{-1} are corresponding to the C=O vibrations of the amide group and the primary amine N-H deformation vibrations, respectively [62]. However, those peaks show redshift in the spectrum of EIPH, and the peak for hydroxide stretching mode and the —NH stretching mode becomes wider in EIPH, compared with PAAm hydrogel, which indicates the formation of hydrogen bonding in the EIPH [63].

In Fig. 1c, the EIPH can be flexibly bent or highly pressed and then recovers to its original shape. The EIPH is therefore an ideal material for capacitive pressure sensors because of its superior elastic property as well as good ionic conductivity [56,64,65]. The mechanical properties of the EIPH were also measured, as shown in Fig. 1d and Fig. 1e. The compression limit is an important parameter, which determines the pressure sensing range of the fabricated sensor. Fig. 1d shows the tests of elastic compression ability of EIPH. The stress shows nearly linear increase with strain up to 55.64% and a jump with still pressing, which indicates the elastic compression of EIPH is 55.64%. Furthermore, it can be seen in Fig. 1e, the strain response of EIPH to pressure is reversible and only a little hysteresis between the compression and relaxation traces is observed in five cycles with 10% strain increasing in each cycle, suggesting the great capability of the EIPH for pressure sensing. Also, the compression Young's modulus can be calculated as 90.14 kPa and 329.2 kPa in the strain range of 0–10% and 20–40%, respectively.

Since microstructure plays an important role in elastomer-based pressure sensors [5,18,28,59], the photopatternable property of the EIPH on a flexible substrate of PET coated with indium-tin oxide (ITO) layer was investigated by using an in-house UV digital lithography technology [59,60]. Fig. 2 shows the laser scanning confocal images of the patterned EIPH micropillar structures with widths of 10, 20, 40, and 80 μm. The fabricated EIPH micropillars are regular and uniform across the whole flexible substrate. A uniform structure is essential for the large-area capacitive pressure sensor to provide a clearly defined and reproducible top contact plane [5]. The thicknesses of the hydrogel films are around 45 μm, and more geometric details of the microstructures are given in Fig. S2. SEM images are also provided in Fig. S3. Notably, the EIPH micropillar structures were quickly optically patterned in 40 s. Therefore, such a photopatternable hydrogel is appealing not only for microscale device fabrication but also for large-scale production.

The micropatterned EIPH on ITO/PET was then laminated with another flexible ITO/PET substrate to form a capacitive pressure sensor of parallel-plate configuration. The schematic of such an EIPH-based pressure sensor is shown in Fig. 3a. When a voltage is applied, an EDL is generated in the EIPH, which gives rise to the high capacitance of the sensor. With a pressure applied on the sensor, the micropillar EIPH structures deform, which causes an increase in the contact area between the electrodes [5,18,59]. As a result, the capacitance of the sensor will increase in accordance with the applied pressure as more charges accumulating at the interface between the electrode and hydrogel [35,36,47,66]. In the experiments, four microstructured EIPH-based capacitive pressure sensors were fabricated and are referred to as sensor-1, sensor-2, sensor-3, and sensor-4, whose micropillars have widths of 80, 40, 20, and 10 μm, respectively. In addition, a bulk EIPH-based capacitive pressure sensor, denoted by sensor-0, was also prepared for comparison. The sensors were then tested by using a force gauge mounted on a stepping motor, and the measurement frequency of the tests was 1 kHz.

Fig. 3b summarizes the measured responses of the four microstructured EIPH-based capacitive pressure sensors. Three or more parallel samples were tested for each type of devices. Here, the sensitivity (S) is defined as $S=(\Delta C/C_0)/\Delta P$, where ΔC is the relative change in the capacitance, C_0 is the initial capacitance of the sensor, and ΔP is a change in the applied pressure. It can be seen that the responses of the microstructured EIPH-based sensors show similar responsive behaviours; *i.e.,* the sensor reveals a linear response with high sensitivity when the applied pressure is low (not higher than 3–3.5 kPa) and then becomes less sensitive when the pressure is further increased. The high sensitivity in the low-pressure range is due to the microstructurally enhanced elastic deformation, which substantially increases the contact

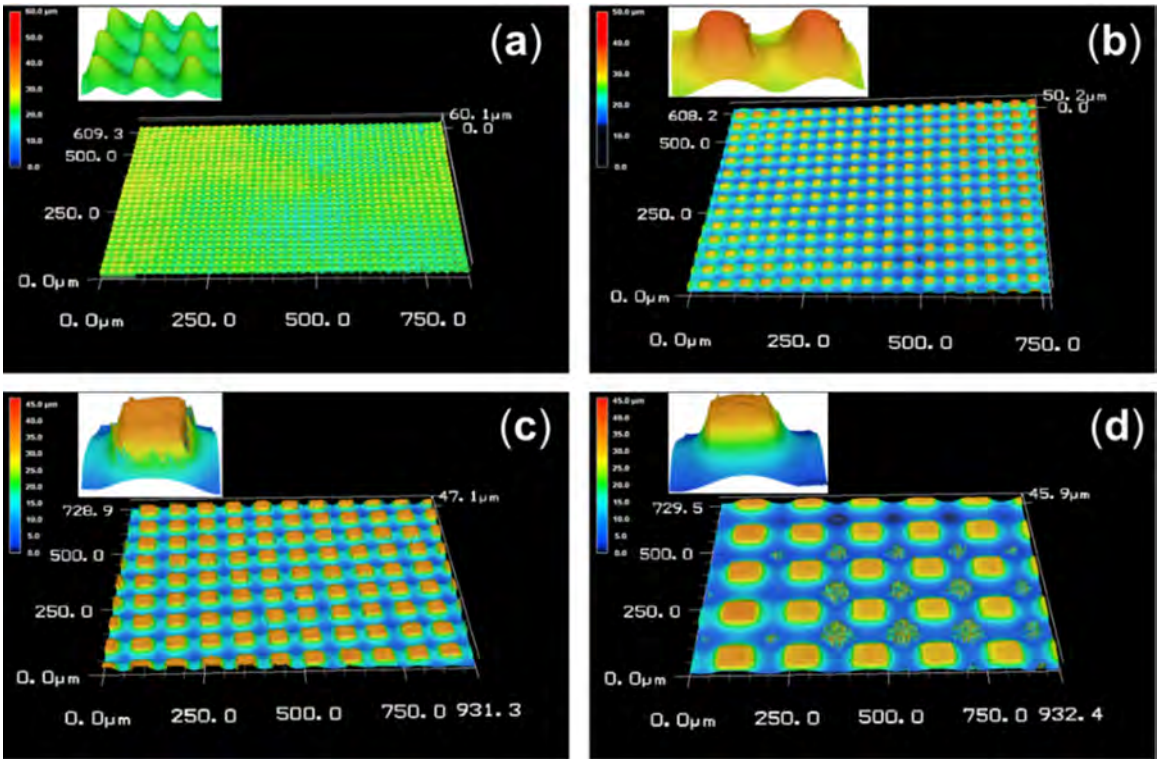

Fig. 2. Laser scanning confocal images of the fabricated EIPH micropillars with different sizes: (a) 10 μm; (b) 20 μm; (c) 40 μm; and (d) 80 μm. The inset shows the enlarged microstructures (both the size and pitch are the same).

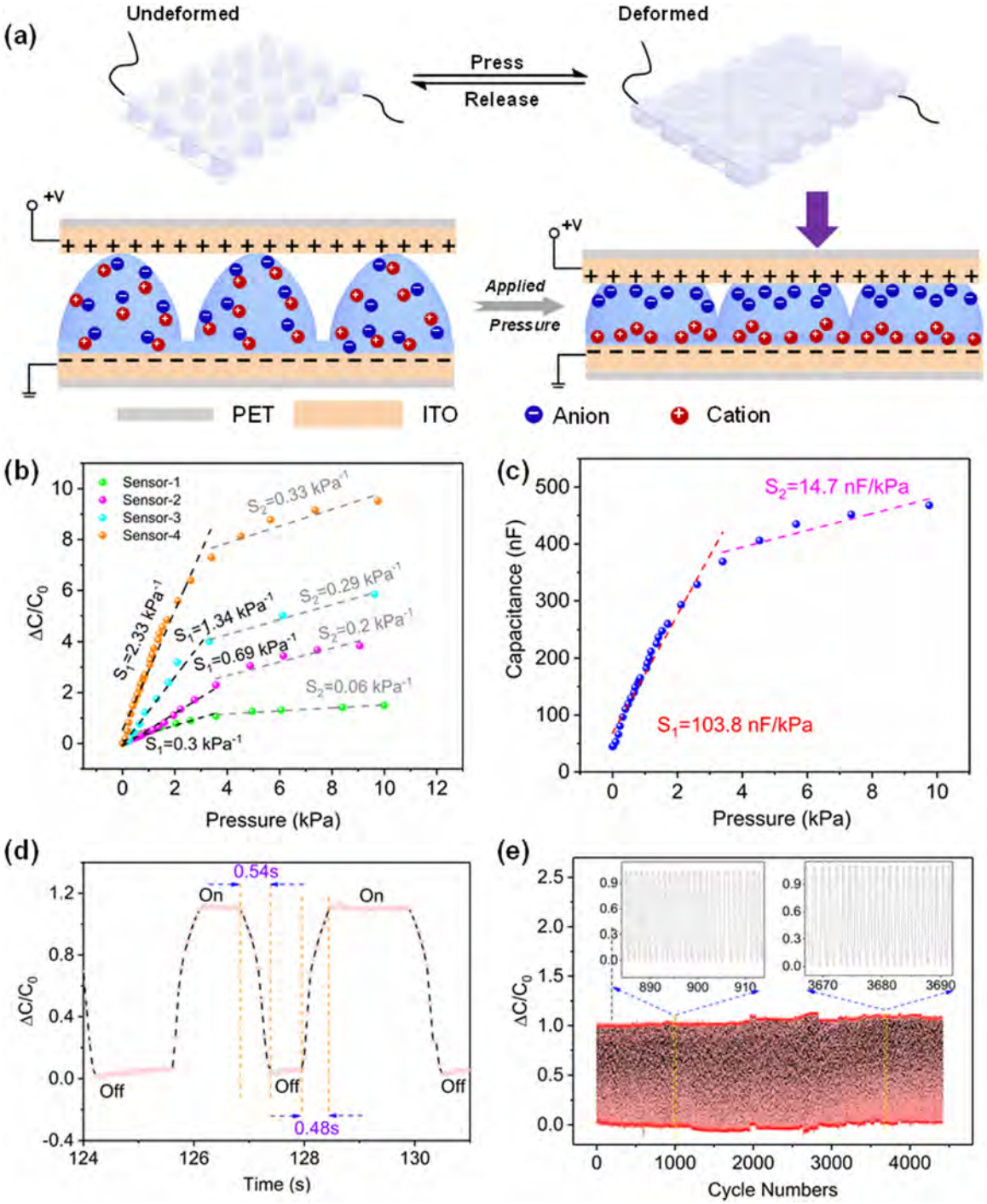

Fig. 3. Schematic and performance of the EIPH-based capacitive pressure sensor: (a) Schematic of the micropatterned EIPH-based capacitive pressure sensor. The top is the shape of the microstructure change with pressure; the bottom is the charge distribution variation in the different states. (b) Pressure-response curves of the pressure sensors with different EIPH microstructures. (c) Capacitance change in the sensor-4 with respect to the applied pressure. (d) Response time of the sensor-4 when the load pressure is 500 Pa. (e) Long-term stability testing result of the sensor-4 under loading/unloading at a pressure of 500 Pa for more than 4000 cycles.

area between the EIPH and ITO electrodes and thereby allows more ions in the EIPH to accumulate around the interface, resulting in a high capacitance [35,36]. For the high-pressure range, the elastic resistance of the EIPH increases because of the breaking of ionic bonds and hydrogen bonds as well as the decrease in the microstructure-induced deformation, which consequently leads to a lower sensitivity [5,18,59].

Notably, sensor-4 shows the highest sensitivity of 2.33 kPa^{-1} with a detection range up to 3 kPa, which is among top-quality capacitive pressure sensors. The capacitance of sensor-4 as a function of the applied pressure is provided in Fig. 3c, while the capacitance responses of all other pressure sensors (with micropillar structures) are given in Fig. S4. It can be seen that the sensitivity of sensor-4 is 103.8 nF/kPa in the pressure range of 0–3 kPa, which is hundreds of times higher than that of conventional capacitive sensors [18,36]. The high-capacitance response is attributed to the large number of ions in the polymer matrix which greatly enhance the capacitance of dielectric layer to nF level (Fig. S4). Moreover, the microstructures enable large change of contact area which further amplify the change of capacitance under pressure. One advantage of the pressure sensor is that it can suppress the adverse influence of the body capacitance (up to hundreds of pF) [35] and other

parasitic sources of noise and considerably enhance the measurement accuracy. Besides, the large capacitance would also reduce the operation voltage of OTFT device, lowering the energy consumption of OTFT-based pressure sensors [48,53]. It indicates the great promise of the sensor for wearable devices applications.

Moreover, the dynamic response of sensor-4 was measured by rapidly loading/unloading a pressure of 500 Pa. In Fig. 3d, the response and relaxation time of sensor-4 are similar: 0.54 and 0.48 s, respectively. The relatively slow response can be attributed to the viscoelastic property of the double-network hydrogel [59,67]. It can be anticipated that the response of the sensor will further slow down a little bit in the high-pressure range because of the increased elastic resistance of the EIPH. Although the response is not fast, it can still be sufficient in many practical applications, such as detection of human motion [68], measuring wrist pulse of human [69], E-hand skin [59], and so on. The durability of the pressure sensor was tested with repeated loading/unloading at a pressure of 500 Pa for more than 4000 cycles, as shown in Fig. 3e. The results indicate that the sensor is stable and shows very little change during the long-time tests.

The effect of bending on the performance of Sensor-4 was also investigated. As shown in Fig. S5, the capacitance of Sensor-4 decreases with the reduction of bending radius. The phenomenon may result from the increase of the distance between microstructures with bending, which reduces the contact area. Notably, the capacitance of the sensor can almost be recovered when the bending radius is not very small (bending radius > 6 mm). Even if the bending radius is as small as 1 mm, the capacitance of the sensor can be recovered to 70% of its original value without bending. The result indicates the good potential of the sensor for wearable devices applications. With the decrease of bending radius, the sensitivity of the pressure sensor decreases a bit in the low-pressure range but increases slightly in the high-pressure range, and meanwhile the range of high-sensitivity region increased a little (Fig. S5c-e). It results from the tilt of micropillars which may require higher pressure to reach the same contact area change and allow more space for deformation under press, as illustrated in Fig. S5b.

Besides, the influence of humidity on the capacitance of the sensor was measured, as shown in Fig. S6. The capacitance of the sensor decreased with the increase of humidity, which appeared a nearly linear variation with respect to the humidity ranging from 35% to 70%. It may result from swelling of the EIPH with the increase of humidity. The swelling of the EIPH led to the thickness increase, which thus reduced the capacitance of the sensor. Moreover, the distance between micropillars may decrease with the swelling of the EIPH, which thereby reduces the sensitivity of the sensor [5,18,59]. Moreover, the swelling degree of EIPH increases with the temperature in a normal temperature region [70], which thus results in the decrease of sensitivity with the increase of environmental temperature.

With its high capacitance and distinguished elasticity, the micropatterned EIPH was thus further adopted as a pressure-sensitive dielectric layer for OTFT-based pressure sensors. Fig. 4a shows the layout of the OTFT-based pressure sensor. In the device, Au source/drain electrodes were deposited on a PET substrate on which an organic semiconductor (OSC) layer (two kinds of OSCs were used here, as shown in Fig. 4b) was deposited *via* spin-coating. Another part employed an ITO/PET film as a gate electrode, on which EIPH was micropatterned. The two parts were then laminated to form the OTFT sensor.

Fig. 4c and Fig. 4d show the source-drain current (I_{ds}) responses of two p-type OTFT devices, whose OSC layers are PTB7-Th and PIDT-BT: TCNQ, respectively, to the external pressure. The sensitivity of the OTFT-based pressure sensors is defined as $S = (\Delta I/I_0)/\Delta P$, where ΔI is the relative change in the current and I_0 is the initial current of the sensor without pressure loading. Importantly, the sensitivities of the OTFT pressure sensors based on PTB7-Th and PIDT-BT: TCNQ are dramatically enhanced to 12.64 and 17.95 kPa^{-1}, respectively. The improvement in the sensitivities originates from the signal amplification function of the OTFT [37]. The I_{ds} of OTFT device in saturation region can be calculated from the equation [53]:

$$I_{ds} = \frac{C\mu W}{2L}(V_{gs} - V_{th})^2,$$

where I_{ds} is the source-drain current in the saturated region, C is the capacitance of the gate dielectric layer, μ is the field effect mobility, W and L are the semiconductor channel width and length, respectively, V_{gs} and V_{th} are the gate voltage and the threshold voltage, respectively. Thus, it can be clearly seen that the value of I_{ds} is proportional to the capacitance of the gate dielectric layer and hence depends on the applied pressure [5,18]. Besides, with the sensitivity of 17.95 kPa^{-1}, the minimum detectable pressure can be estimated to be 0.18 Pa at a signal-to-noise ratio of 3. The remarkably high sensitivity of OTFT pressure sensors may resort to two respects: 1) micropatterned EIPH leads to the large change of capacitance of dielectric layer; 2) the formation of EDL which depends on the contact area of dielectric layer and semiconductor layer remarkably modulates the hole generation of semiconductor channel [53]. Notably, the sensitivities of the two OTFT-based pressure sensors are not the same, which can be attributed to the hole mobility variation with the change between the dielectric layer and semiconductor [71]. In addition, a similar phenomenon was also observed in previous OTFT-based pressure sensors [19]. Moreover, the output I_{ds} closely depends on the hole mobility of the OSC layer. It can be seen I_{ds} in Fig. 4e is one order of magnitude larger than that in Fig. 4f, which indicates the much higher mobility of PTB7-Th than that of PIDT-BT: TCNQ (the output curves of two OTFT-based pressure sensors with and without applying pressures were supplied in Fig. S8).

Table 1 summarizes our results as well as some similar OTFT-based pressure sensors whose dielectric layer is an elastic polymer. Compared to other similar OTFT-based pressure sensors, the pressure sensors in our study not only exhibit higher sensitivities but also demonstrate a much lower operation voltage. Our OTFT-based pressure sensors can operate stably at the bias voltages as low as 2 V due to the large capacitance of the EIPH [35,36,66], which makes them promising for integration into various microdevices for portable and wearable electronics. In addition, compared with previous OTFT-based pressure sensors whose operating voltages are tens or hundreds of volts, the operating voltage in this work is much lower (Table 1) [5,18,19,42].

To demonstrate the ability of the EIPH-based pressure sensors for tactile sensing applications, we fabricated a proof-of-concept capacitive matrix-type pressure sensor by laminating the micropatterned EIPH-coated ITO/PET substrate with a 4 × 4 electrode array, as shown in Fig. 5a. The area of each pixel is 49 mm^2. Three objects, made from polystyrene, of different characters such as 'O' and 'K' were used for testing. The objects were placed on the top of the sensor arrays and the induced change in capacitance at each pixel was recorded. Fig. 5(b-d) show the measured response of the sensor array to the placement of these objects, where a greener color represents a larger capacitance. The spillover of the signals from an addressed pixel into a neighboring pixel can be seen and can be improved using smaller pixels. The differences in the color intensity originated from the uneven surface, *i.e.,* the texture structure, of the object. The sensor array shows great potential for multitouch devices.

It is known that microstructures play a key role in improving the sensitivities of pressure sensors [5,18,50,59]. Indeed, compared with the response of the sensor-0 based on the EIPH film without microstructures, as shown in Fig. S9, the sensitivities of pressure sensors with micropillar structures are much higher. In general, the improved sensitivity is due to the micropillar structures that provide more spaces for hydrogel deformation, *i.e.,* the micropillar structures reduce the elastic resistance [5]. Moreover, the geometrical parameters of the microstructures also considerably influence the performance of the pressure sensors. In Fig. 3b., the sensitivities of the microstructured EIPH pressure sensors decrease with an increase in micropillar size. This behavior can be attributed to the following two factors: 1) the effect of smaller

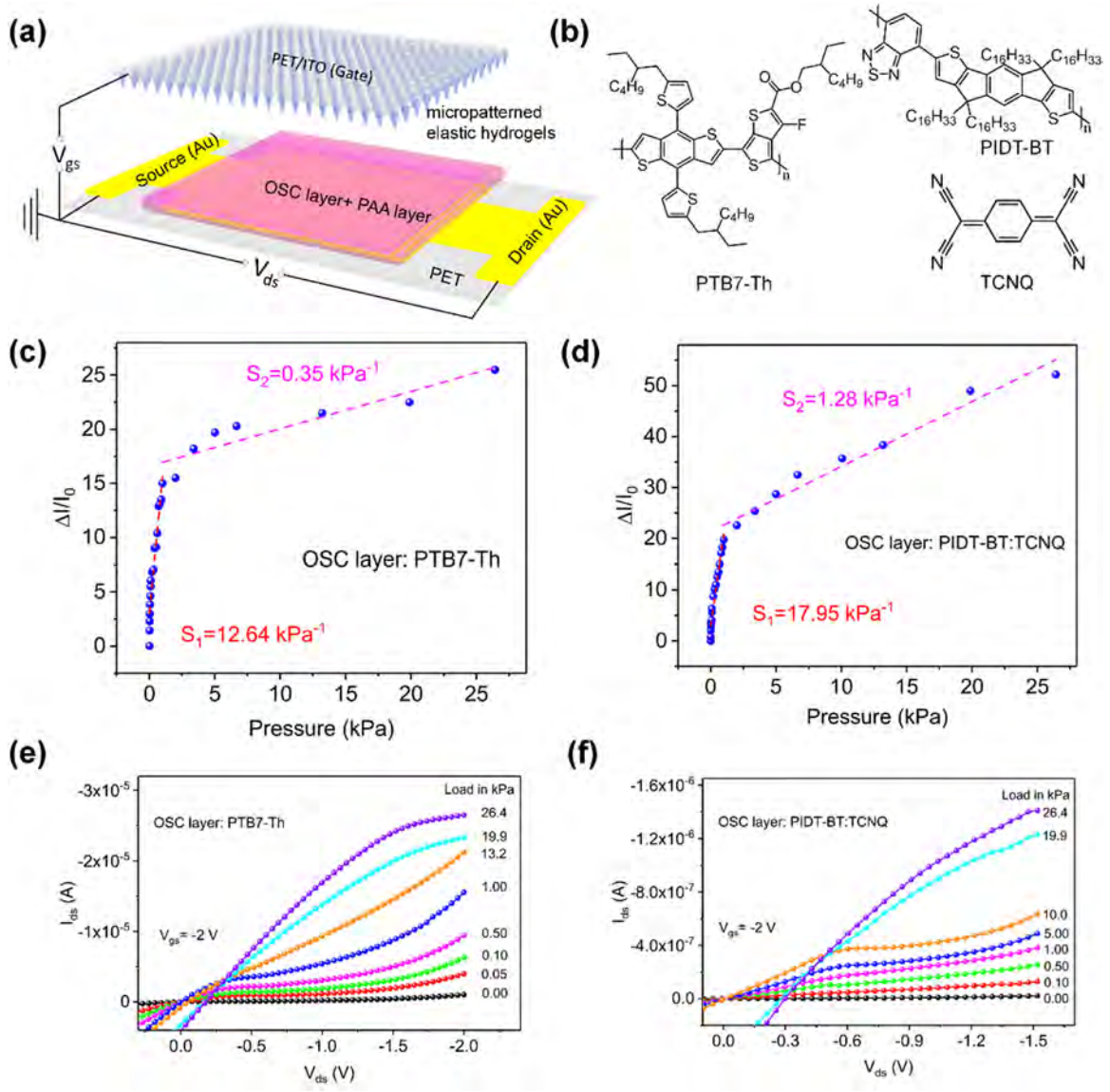

Fig. 4. Schematic and performance of the EIPH-based OTFT pressure sensors: (a) Schematic of the EIPH-based OTFT, consisting of polymer organic semiconductors and a micropatterned EIPH dielectric layer; (b) chemical structures of the organic semiconductors used in OTFT; (c) and (d) saturated current responses of the two OTFT pressure sensors with the OSC layers of PTB7-Th and PIDT-BT: TCNQ; (e) and (f) are the corresponding output drain currents under different pressures.

micropillars is more significant in the microstructure-enhanced elastic deformation; 2) the contact area between the upper electrode and EIPH hydrogel also depends on the sizes of micropillars, which determines the variation in the capacitance. The former effect has been reported previously [5]. The latter effect can be proven by estimating the initial contact area of the four sensors from Fig. S2. The results are summarized in Fig. S10, which are 0.16, 0.21, 0.31, and 0.42 mm^2 for sensor-4, sensor-3, sensor-2, and sensor-1, respectively, within the fixed probe area of 1.77 mm^2. It can be logically interfered that the smaller initial contact area will provide much more space change. With an applied

Table 1
Comparison of the present pressure sensor with previously reported elastic polymer-based capacitive sensors.

Types of transduction	Materials	Sensitivity (kPa^{-1}) (Sensitivity region)	Operating voltage (V)	Ref.
Capacitance	Ecoflex (porous)	0.601 (0–5 kPa)	–	[17]
Capacitance	PDMS (porous)	0.26 (0–0.33 kPa)	–	[16]
Capacitance	PDMS (microstructure)	0.55 (0–2 kPa)	–	[5]
Capacitance	PDMS (microstructure)	0.76 (0–2 kPa)	–	[28]
Capacitance	PDMS	0.00023 (0–900 kPa)	–	[11]
Capacitance	PAAm (microstructure)	2.33 (0–3 kPa)	–	This work
OTFT	PDMS (microstructure) /PiI2T-Si	8.2 (0–8 kPa)	200	[18]
OTFT	Polyurethane (nanostructure)	1.76 (0–1 kPa)	50	[72]
OTFT	(PMMA/PAA)/PIDT-BT: TCNQ	56.15 (0–5 kPa)	5	[48]
OTFT	PAAm (microstructure) /PTB7-Th	12.64 (0–1.5 kPa)	2	This work
OTFT	PAAm (microstructure) /PIDT-BT: TCNQ	17.95 (0–1.5 kPa)	2	This work

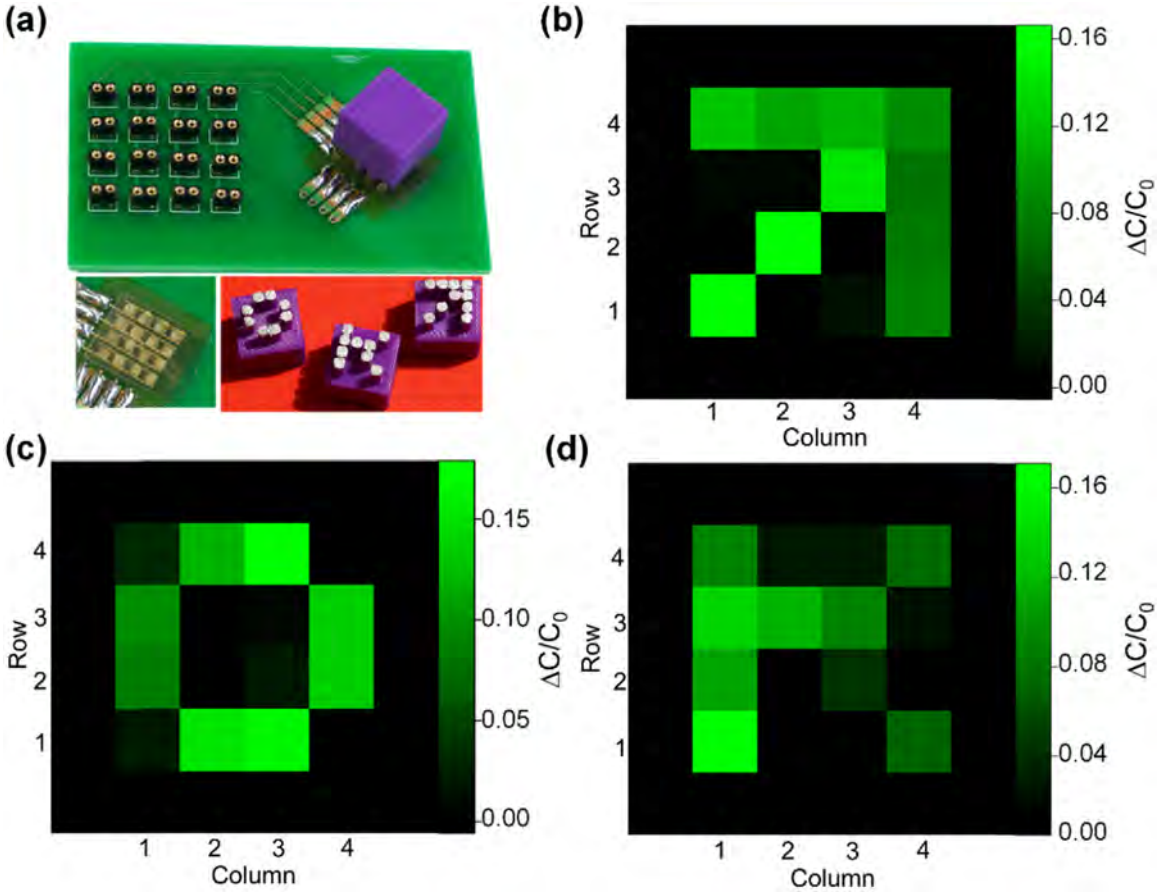

Fig. 5. Demonstration of pressure mapping using the EIPH-based capacitive sensor array: (a) Photos of capacitive matrix-type pressure sensor and the plastic objects with different textures; the response of the device to the placement of objects with different textures: arrow shape (b); O shape (c); and K shape (d).

pressure, a change in contact area leads to a dramatic change in capacitance as the capacitance of an EDL structure depends on the contact area with electrodes [35,49,50].

4. Conclusion

In summary, we designed and prepared a novel type of elastic ionic hydrogel with high capacitance, *i.e.,* EIPH, by photopolymerization of an acrylamide monomer in an aqueous solution of poly (acrylic acid) and $CaCl_2$. The EIPH was *in situ* micropatterned on an ITO electrode and laminated with another electrode to fabricate capacitive pressure sensors. Because the high elasticity and capacitance of these sensors, the capacitance sensitivity achieved in this work is more than 100 times larger than that of conventional capacitive pressure sensors, which suggests top-quality capacitive pressure sensors. With the pressure-sensitive property and high capacitance, the micropatterned EIPH was adopted for the first time as a dielectric layer for the fabrication of OTFT-based pressure sensors, which further improved the sensitivity to 17.95 kPa^{-1} with an operation voltage as low as 2 V. The high sensitivity and low-power operation of these flexible capacitive and OTFT pressure sensors render them promising for applications in wearable devices.

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Appendix A. Supporting information

Supplementary data associated with this article can be found in the online version at doi:10.1016/j.nanoen.2019.01.032

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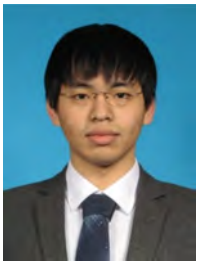

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