
A 1.3-micrometre-thick elastic conductor for seamless on-skin and implantable sensors

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Supplementary Table 2: A comparison of the ultrathin neural electrode with previously reported cuff-type flexible and stretchable nerve electrodes.

✓ **Captions for the Supplementary Videos**

✓ **Supplementary references**

Experimental Section:

Device characterisation and analysis

For the mechanical property test, the ultrathin PDMS-Au conductor was horizontally fixed to the stage of a high-precision mechanical system (Thorlabs model LTS150/M, USA), where a uniform stretch/release test was controlled using a computer-based user interface (Thorlabs APT user). During the mechanical test, the resistance of the ultrathin PDMS-Au conductor was recorded in real time using a digital multimeter (34410A Agilent, USA)—the sample size being kept the same: a width of 5 mm and length of 5 mm, five samples of which were evaluated for each condition. SEM images were taken using an S4800 (Hitachi High-Technologies) with an accelerating voltage of 5 kV. For measuring the resistance change of the ultrathin PDMS-Au conductor when attached to human skin, we utilized a digital multimeter (34410A Agilent, USA) to connect two copper wires, which connected the ultrathin PDMS-Au conductor via liquid metals. To attach the ultrathin PDMS-Au conductor to human skin, we first pre-stretched the skin by deforming the joint part, and then attached the ultrathin PDMS-Au conductor. After releasing the strain, the ultrathin PDMS-Au conductor formed a skin-mimic texture structure, similar to a buckling structure.

For the adhesion measurement, a hollow frame with a window of $4 \times 4 \text{ cm}^2$ was used as the holder to support the films. All films were fixed onto the holder via double side tape. The initial adhesion area was $\sim 3.6 \text{ cm}^2$ between the suspended films and the skin. The adhesion force was recorded using a high-precision tensile tester (MTS C42) when the holder was gradually lifted at 0.1 mm/s until they completely separated. For the in-water measurement, the contact between the films and the skin was formed firstly, and then water was added into the glass beaker. After a certain

time, the adhesive force was recorded at the same condition.

For the bonding strength characterization, we coated the thin PDMS layer onto a 15 μm -thick polyimide film to make sure that the joint fails before the PDMS layer itself (**Supplementary Fig. 39**). For the PDMS-Au samples, we patterned the Au using the same shadow mask for the fabrication of ultrathin neural electrodes. Then we treated the surfaces of the samples using an oxygen plasma treatment at a power of 30 W for 18 s, and we bonded two samples by adhering them together with a gentle pressure of around 1 kPa. Then the bonding force was recorded using a high-precision tensile tester (MTS C42) when the holder was gradually lifted at 0.1 mm/s until they completely separated.

For the ECG recording, a Heal Force PC-80B ECG Monitor with three electrodes was used.¹ Two electrodes were pasted on each forearm to detect ECG signals, and one reference electrode on the back of the hand. Commercial electrodes (Chunfeng YD-50 Ag/AgCl gel electrodes) or the ultrathin PMDS-Au-pDAM electrodes were used. For measuring the ECG signals under extreme conditions, the volunteer conducted the following actions: 1. Washing hands normally with a water speed of 2.0 L/min from a pipe of 6.0 mm diameter; 2. Running at a speed of 6 km/h for 30 min; 3. Swimming at a speed of 1 km/h for 30 min in breaststroke style; 4. Wearing the electrodes for 8 hours while working normally, including changing gloves when entering chemistry lab. The study protocol related to on-skin ECG sensors was thoroughly reviewed and approved by the ethical committee of Nanyang Technological University (Approval No. IRB-2017-08-035-01).

A vibrator (K2007E01) was utilised to supply the periodic pressure for the characterisation of the ultrathin self-powered sensor as a pressure sensor. The pressure was calibrated using a force gauge (SHIMADZU corporation, module: PFG-10KNA).

The same vibrator was utilised to supply the periodic stretch/release deformation for the characterisation of the ultrathin self-powered sensor as a strain sensor. The tensile strain was calibrated by measuring the deformation of the sensor in a real-time recorded video. In the above two cases, the output current signals from the sensor were recorded using a Stanford low-noise current preamplifier (Model SR570) and NI PCI-6259. The study protocol related to ultrathin mechanical sensor was thoroughly reviewed and approved by the ethical committee of RIKEN (Approval No. Wako3 30-32).

An electrochemical workstation (ZENNIUM X, ZAHNER-elektrik) was used for the electrochemical impedance spectroscopy (EIS) measurement, which was performed in 0.01 M phosphate-buffered saline (PBS) solution with conventional three-electrode configuration. The measured frequency range in EIS was from 100 Hz to 1 MHz with a sine wave current of 5 mV in amplitude and direct current potential of 0 V.

In vivo neural modulation/signal-recording

The experiments were performed in four adult male Sprague–Dawley rats (350–600 g). Animal care and related procedures were approved by the IACUC of National University of Singapore. Animals were anesthetized using gas anesthesia (O_2 2 L/min, isoflurane 2%). Carprofen (5 mg/kg) and normal saline (0.2–0.5 mL/10 g) were injected subcutaneously to provide analgesia and prevent dehydration of the animal. After the animal reached an adequate depth of anesthesia, the sciatic nerve at the left side was exposed, and the ultrathin neural electrodes were placed around the proximal segment of the sciatic nerve. For the control experiment, 100 μ m-thick PDMS-Au-PEDOT:CNTs were fabricated with the same procedure for the ultrathin

neural electrode.

For nerve stimulation experiment, electrical stimulation (biphasic waveform, 1Hz) was applied to the sciatic nerve through our developed neural electrodes, and the evoked CMAPs were recorded from TA and GM muscles through two pair of stainless-steel needle electrodes (Natus Neurotechnology). For nerve recording, electrical stimulation (monophasic waveform, 1Hz) was applied to the left paw of the rats through a pair of needle electrodes, and the evoked CNAPs were recorded from the sciatic nerve through our developed neural electrodes. To confirm the CNAPs are the electroneurogram (ENG) signals, instead of other signals like electromyography (EMG) signals, lidocaine was applied to the nerve for blocking the nerve function. No impulses were recorded except for the stimulus artifact under the same electrical stimulus after the blocking (**Supplementary Fig. 46**). The study protocol related to the implantable experiment was thoroughly reviewed and approved by the IACUC of National University of Singapore (Protocol No. R20-1312).

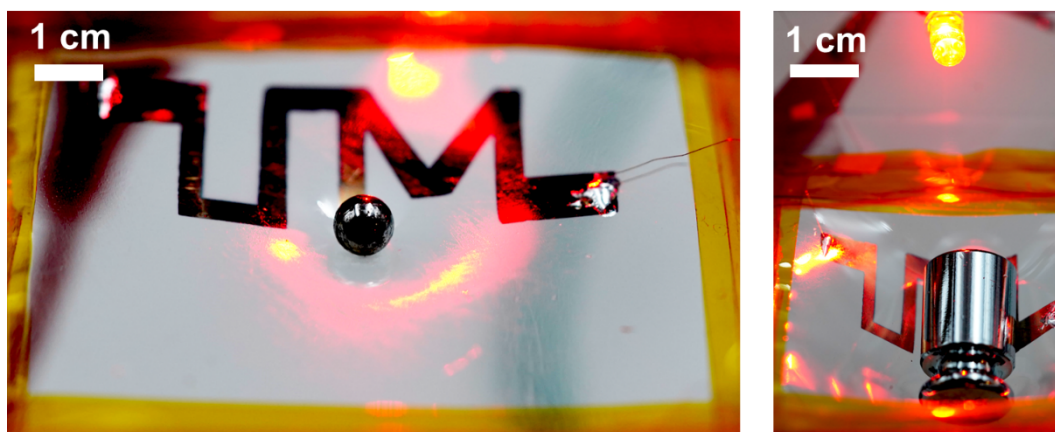
The electrical stimulation was delivered with an isolated pulse stimulator (A-M system, model 2100). The RHD2132 and RHD2216 evaluation system (Intan Technologies) was used for nerve and muscle signal recording, respectively, with a sampling rate of 30 kHz. MATLAB (MathWorks) was employed to process the data. For signal recording under electrical stimulation, the data in one trial which consisted of multiple recordings were averaged (20 recordings for nerve recording experiment and 10 for nerve stimulation experiment), and the signal and noise levels of the averaged waveform were presented. Statistical analysis was conducted in Origin.

COMSOL simulation

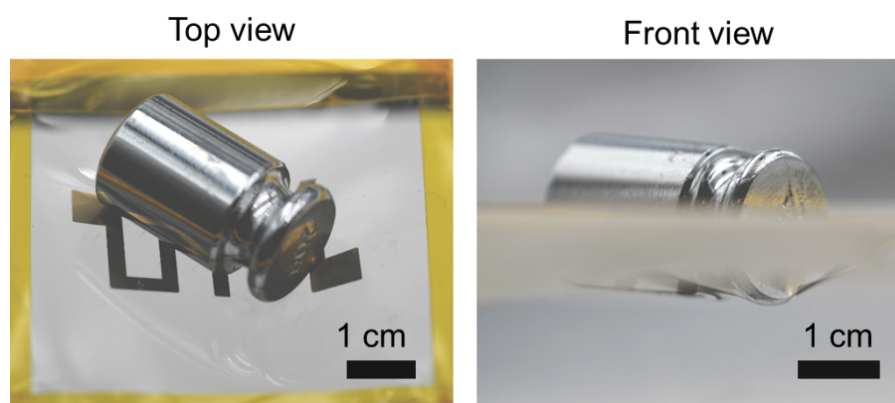
The thermal stress and deformation of the Au layer on the PDMS substrate were

simulated using a commercial finite-element package (COMSOL v5.3, thermal stress). The thickness of the Au layer was fixed at 10 nm, and the shape of the cracks was simplified into different shapes, including Y, T, I, and X. The width of the cracks was fixed at 0.5 μm while the length varied depending on the shape. The dimension of the PDMS substrate was fixed at 400 μm in both the y- and z-directions, and thickness was changed from 100 to 1 μm in the x-direction. A fixed boundary was set for the bottom surface of the PDMS because it was attached to the glass substrate in the experiment, while the other surfaces were set using a free boundary. Considering that only 10 nm-thick Au was deposited on the PDMS substrate, the temperature should not have been higher than 80 $^{\circ}\text{C}$ based on previous reports and our experimental data (**Supplementary Fig. 10**)². Therefore, we set the temperatures to 30, 50, and 80 $^{\circ}\text{C}$ to simulate the steady thermal stress and deformation of the Au layer. The areas of cracks were calculated by a 2D Image Analysis Software (Image-Pro Plus), and 4 cracks were selected for each condition.

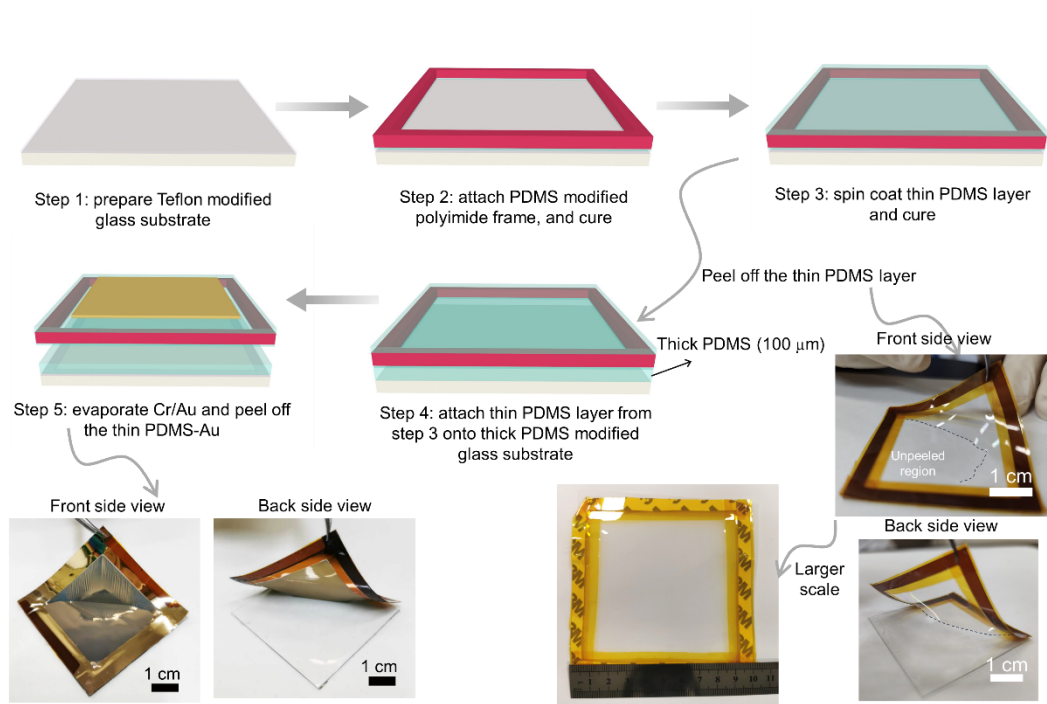
Supplementary Figures



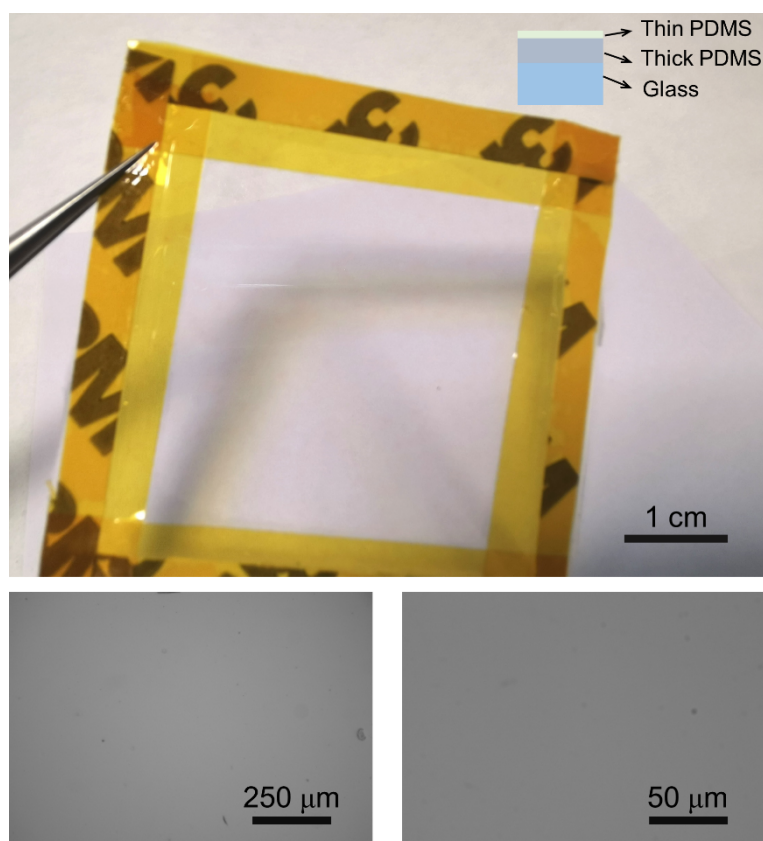
Supplementary Fig. 1: Optical images (top view) of the ultrathin PDMS-Au conductor with a pattern in the shape of “ μm ”, being deformed by different weights. The ultrathin PDMS-Au conductor was connected to a red light-emitting diode (LED) and power source via liquid metal and copper wires.



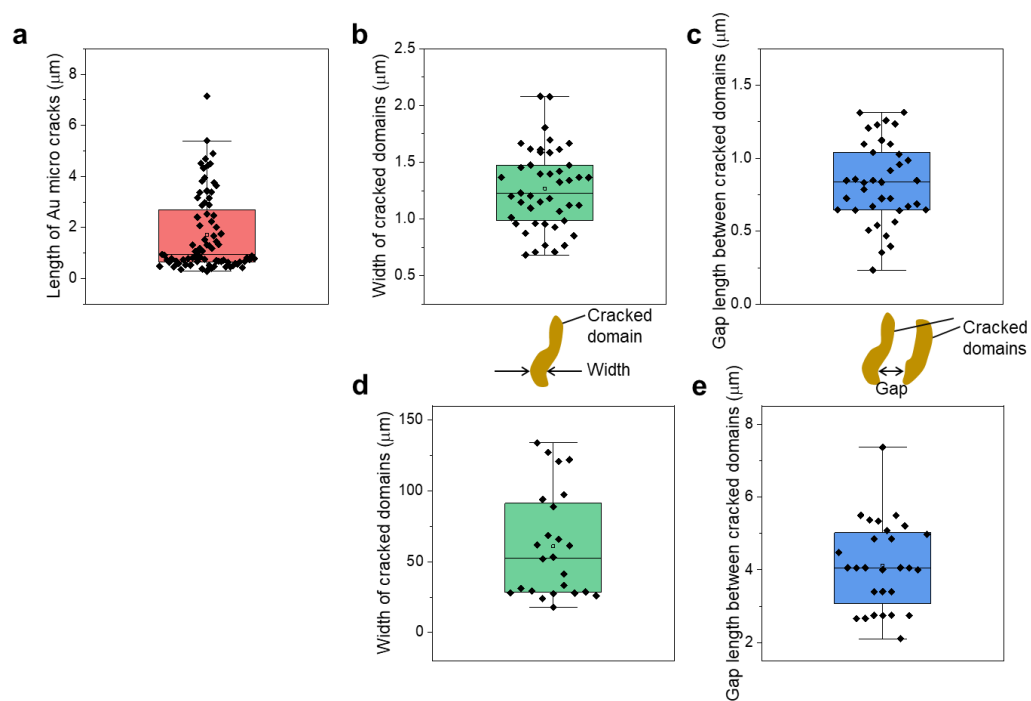
Supplementary Fig. 2: Optical images (top view) of the thick electrode being deformed by a weight of 20 g. It is clear that the deformation of the PDMS much smaller than that of the ultrathin PDMS-Au conductor.



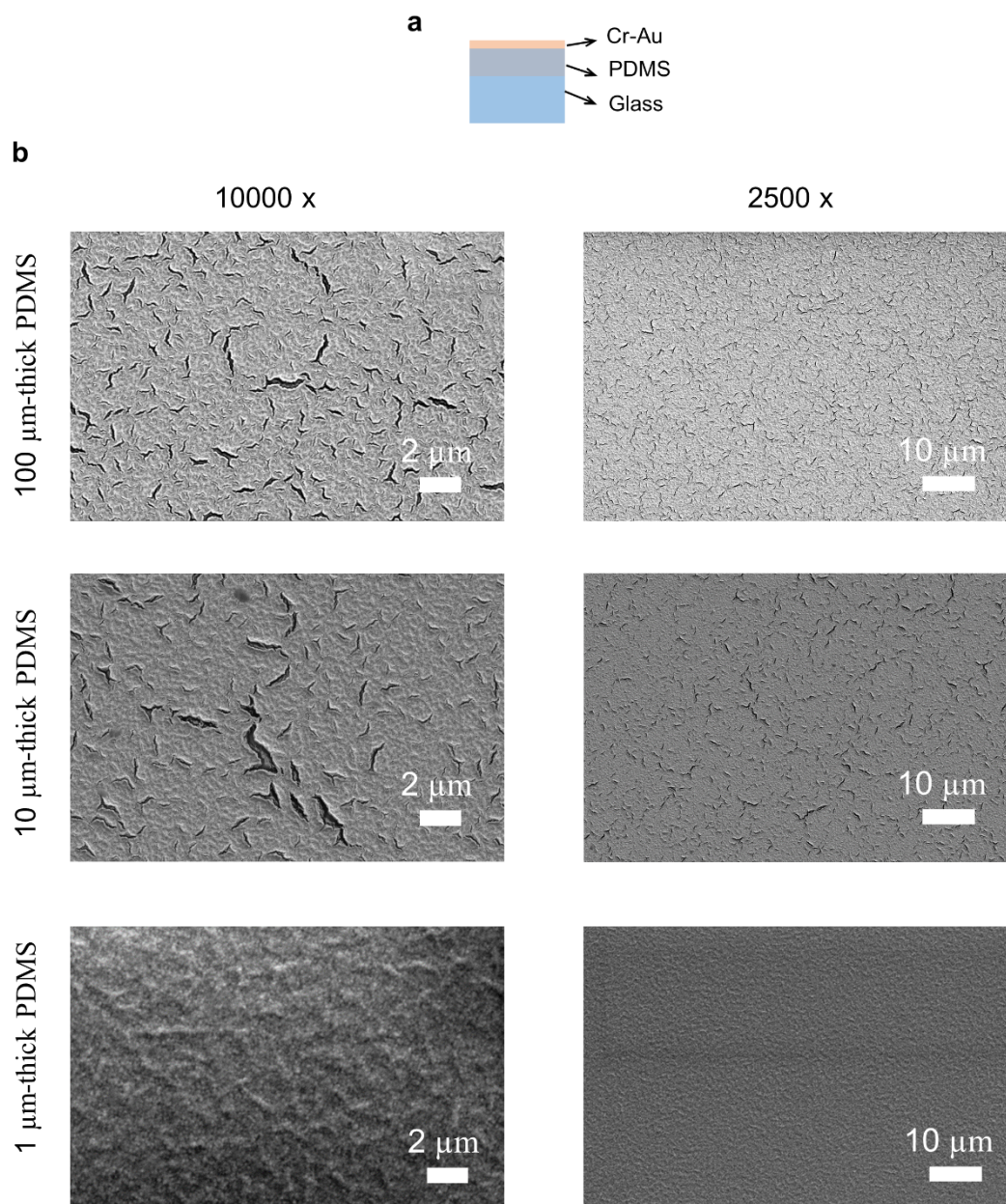
Supplementary Fig. 3: Fabrication process for the ultrathin PDMS-Au conductor.



Supplementary Fig. 4: Optical images of the ultrathin PDMS film transferred onto the thick PDMS coated glass substrate. It is clear that there is no obvious air bubble between the thin and thick PDMS films.

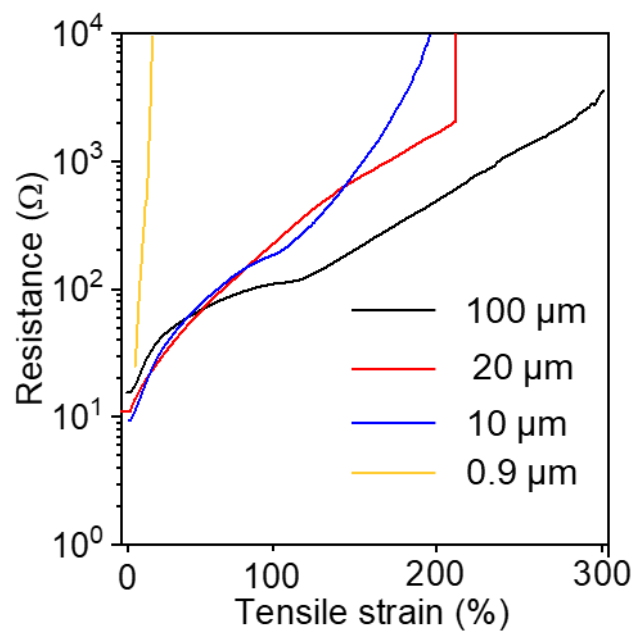


Supplementary Fig. 5. Quantitative analysis of the micro cracks at 0% strain and cracked domains under stretch. a-c. The ultrathin conductor with using our key approach.. d-e. ultrathin PDMS-Au without using our key approach. a. length of Au micro cracks at 0% strain. b,c. Width of cracked domains and gap length between cracked domains at 100% strain. d,e. Width of cracked domains and gap length between cracked domains at 20% strain. It should be noted that for ultrathin PDMS-Au without using our key approach, there was no initial Au micro cracks. (n=3 for all conditions) From bottom to top, the boxplots include the minimum value, first (lower) quartile, median, third (upper) quartile, and maximum value.

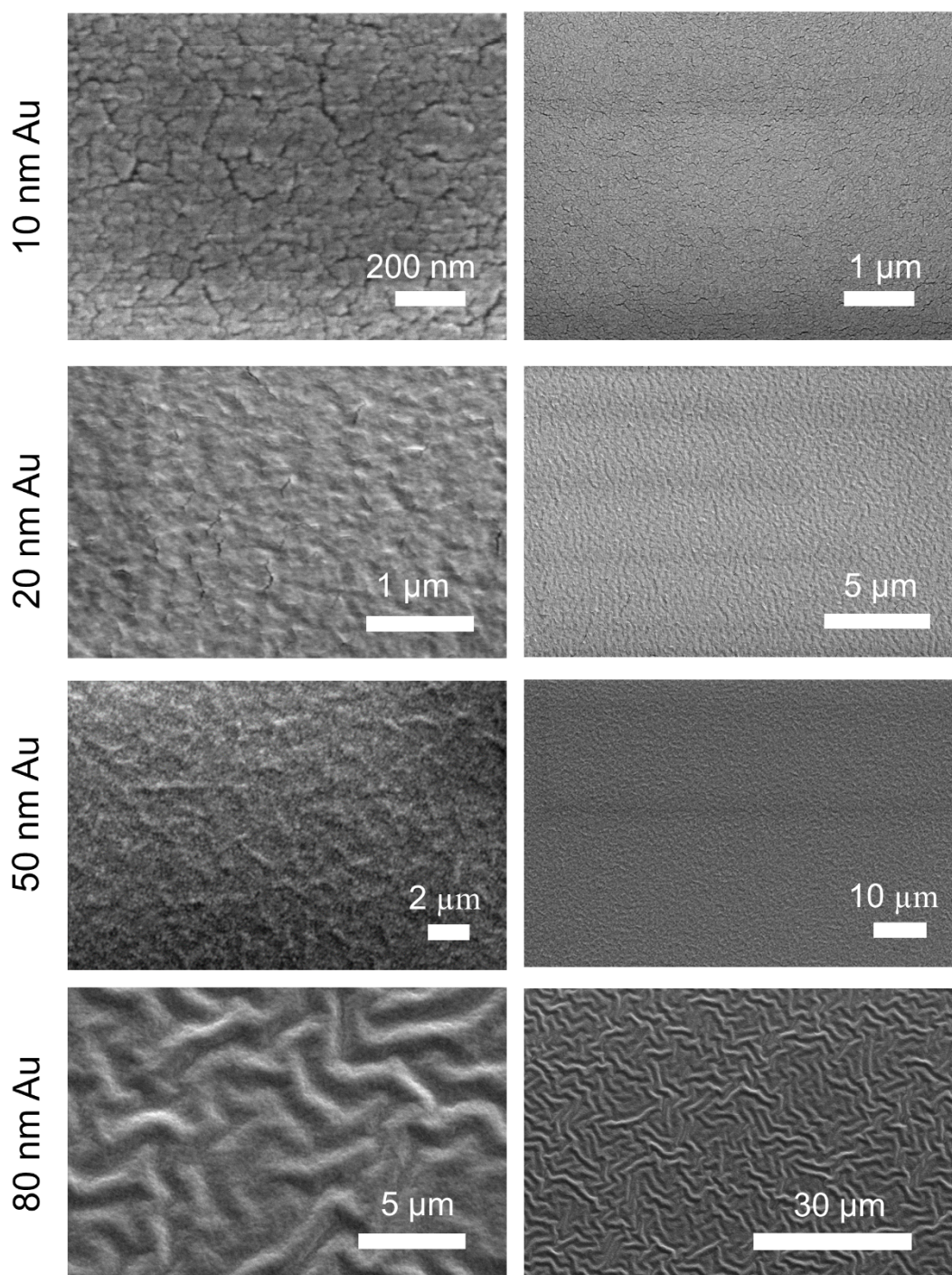


Supplementary Fig. 6: Morphology characterisation of the PDMS-Au conductor with PDMS films of different thicknesses. a, Structure illustration for the fabrication process of the PDMS-Au conductor. b, Comparison of SEM images of the PDMS-Au of different thicknesses. It is clear that the density of the micrometre-scale cracks gradually decreases and reaches zero by decreasing the thickness of the PDMS

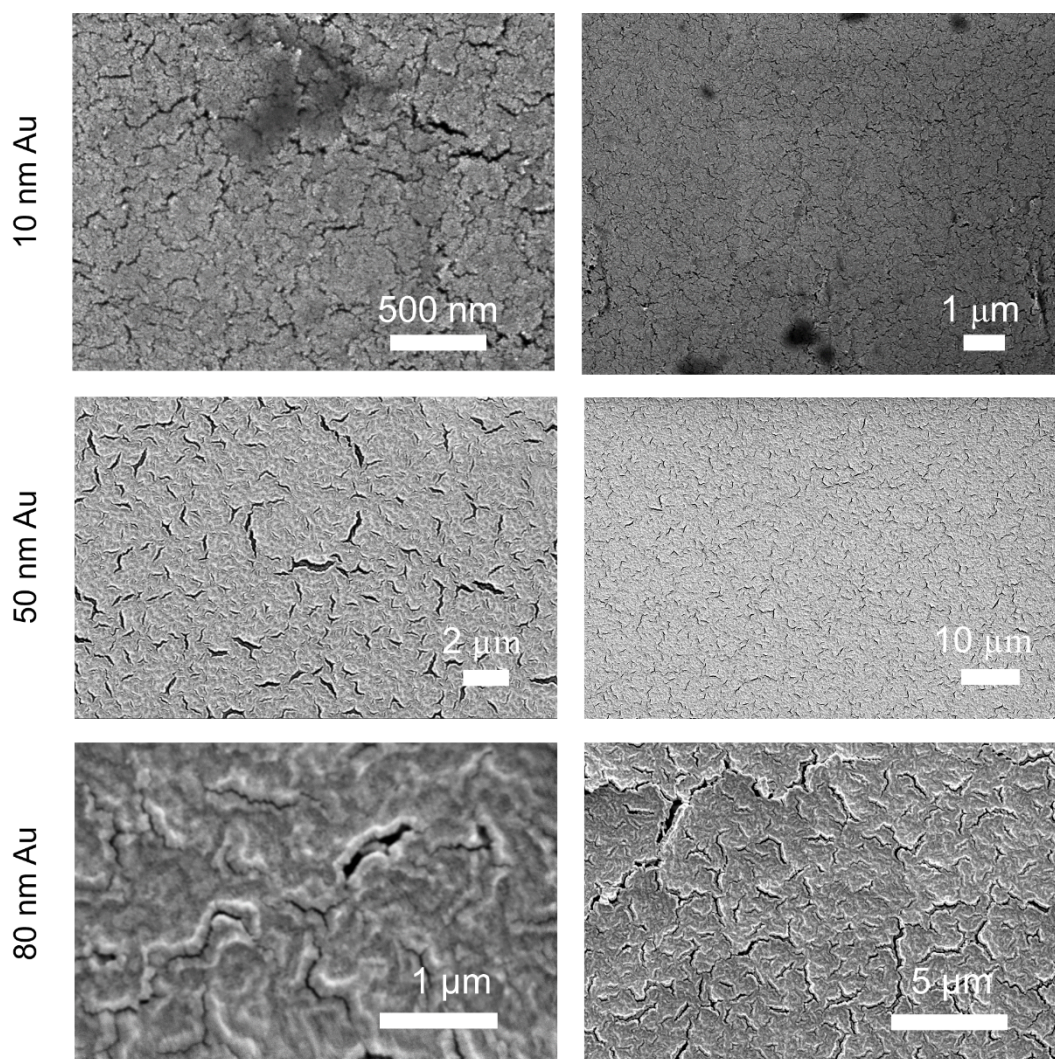
from 100 to 10 to 1 μm . The SEM observation was repeated 3 times.



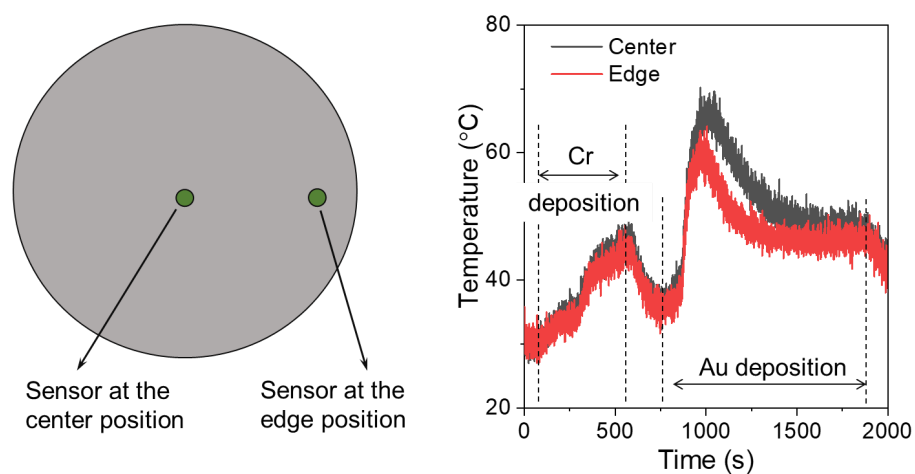
Supplementary Fig. 7: A comparison of the resistance under tensile strains of the PDMS-Au conductor with PDMS films of different thicknesses.



Supplementary Fig. 8: SEM images of the micrometre-thick PDMS-Au conductor with Au layers of different thicknesses using a PDMS film directly spin-coated on the glass substrate. The SEM observation was repeated 3 times.

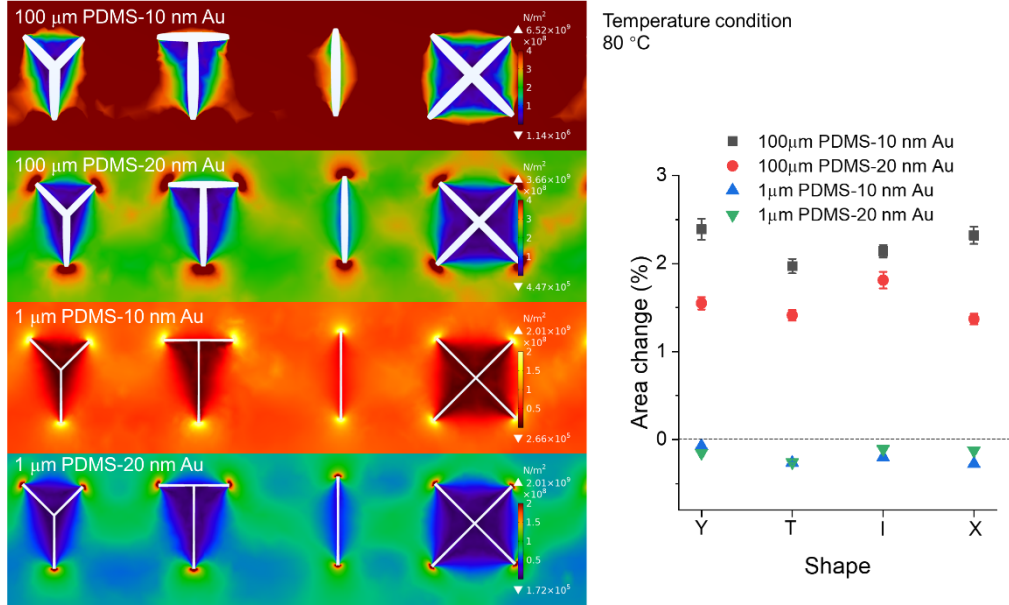


Supplementary Fig. 9: SEM images of the thick conductor with Au layers of different thicknesses. The SEM observation was repeated 3 times.

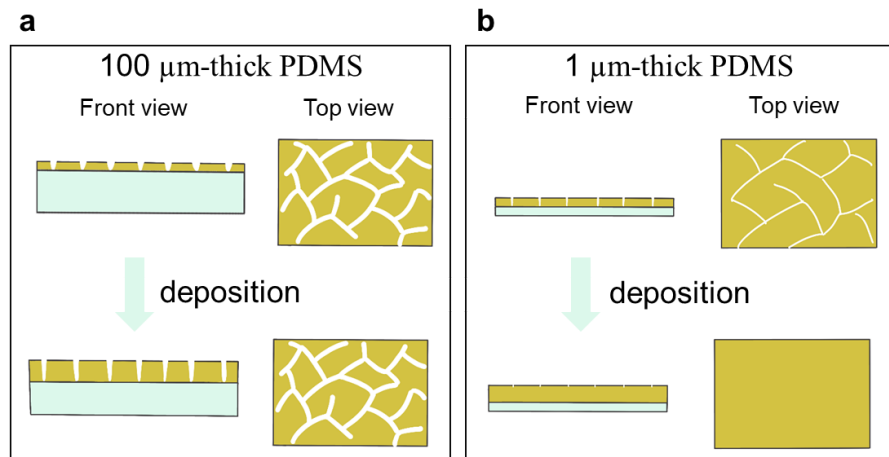


Supplementary Fig. 10: Real-time temperature recording of the samples during

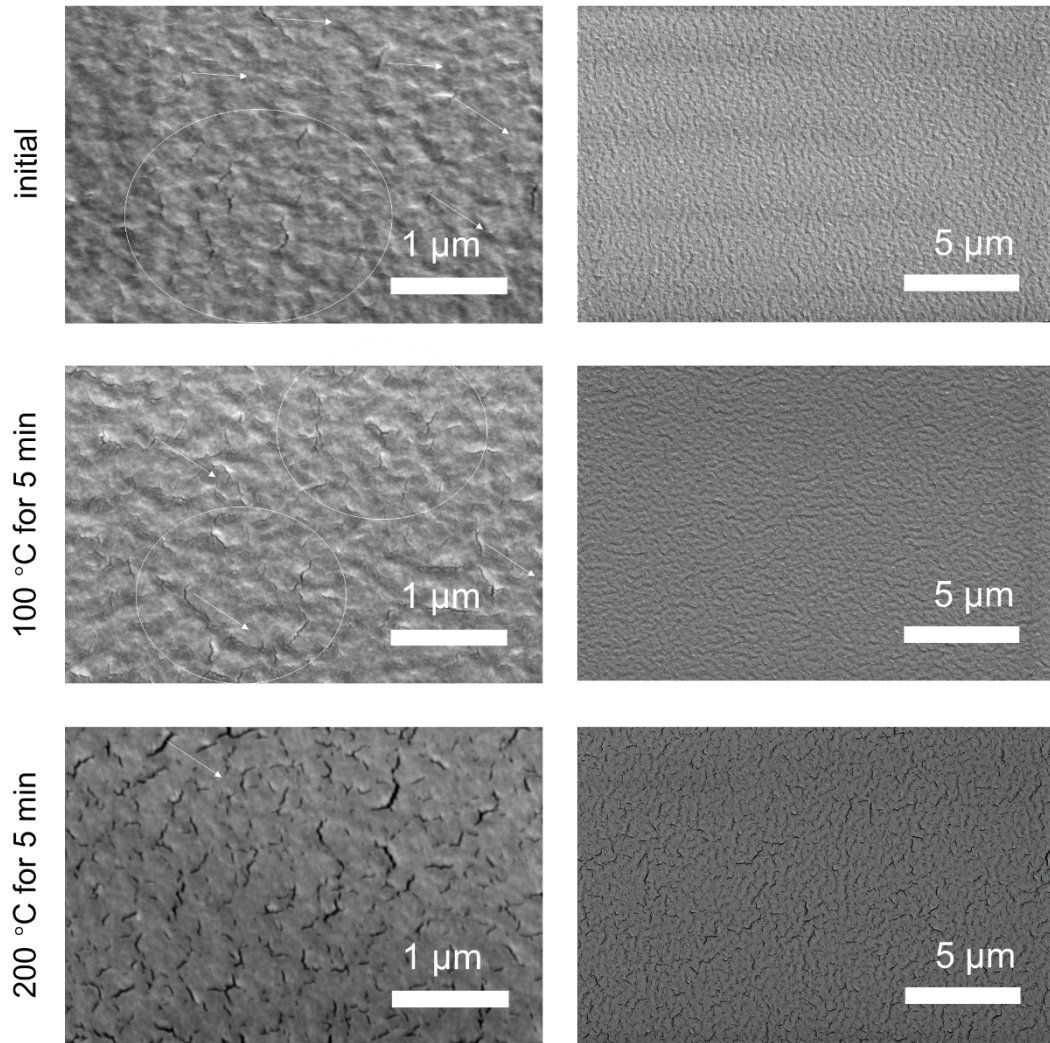
the thermal evaporation process. The temperature on the sensors changed in the range from ~ 30 to 70°C . It should be noted that the instant temperature on the sample can be higher since heat conduction needs time.



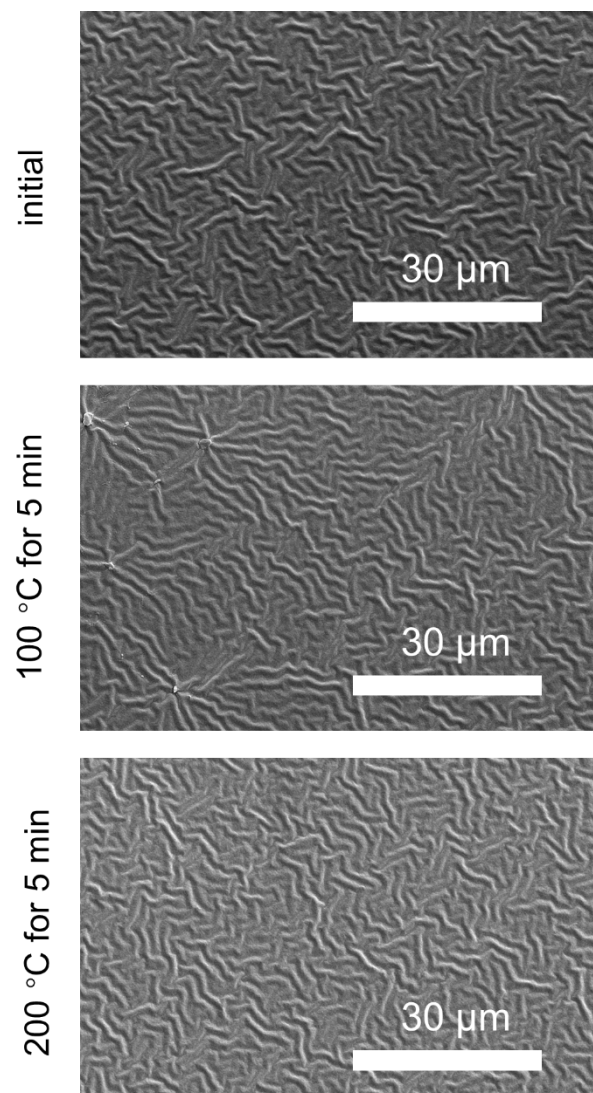
Supplementary Fig. 11: Simulation results for 10 nm- and 20 nm-thick Au on the 1 μm -thick and 100 μm -thick PDMS at 80°C . It should be noted that the deformation was enlarged by 10 times in two directions ($n=4$ for all conditions). Data are presented as mean $\pm$ standard error of the mean (s.e.m.)



Supplementary Fig. 12: Illustration for the Au micro-cracks' growth process on the 100 μm -thick and 1 μm -thick PDMS. a, There is a high density of Au micro-cracks when the thickness of Au is thin. Since the width of the cracks increases with the Au deposition, the micro-cracked morphology is retained when the Au layer is thicker. **b,** There are only a few Au micro-cracks when the thickness of Au is thin. Since the width of the cracks does not change with the Au deposition, the micro-cracked morphology disappeared when the Au layer is thick.

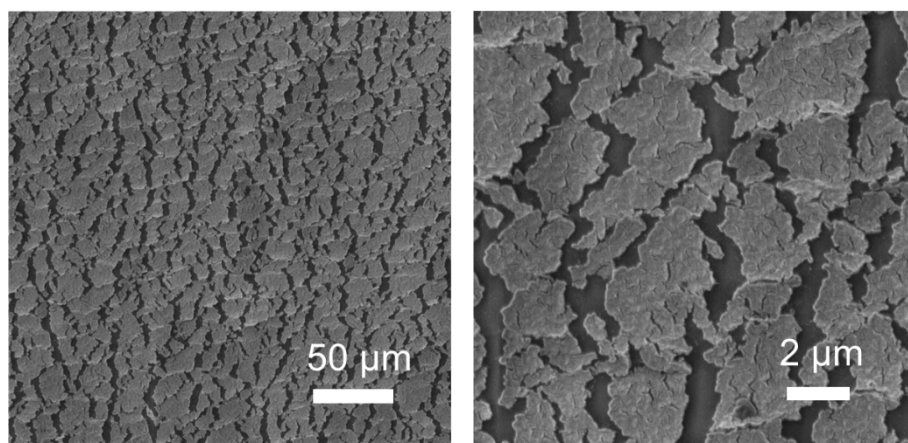


Supplementary Fig. 13: SEM images of the 20 nm-thick Au layer on the 1 μm-thick PDMS substrate without using the key the approach, i.e., the utilization of the 100 μm-thick PDMS between the glass and the 1 μm-thick PDMS substrate, before and after post-annealing at 100 and 200 °C for 5 min. The SEM observation was repeated 3 times.

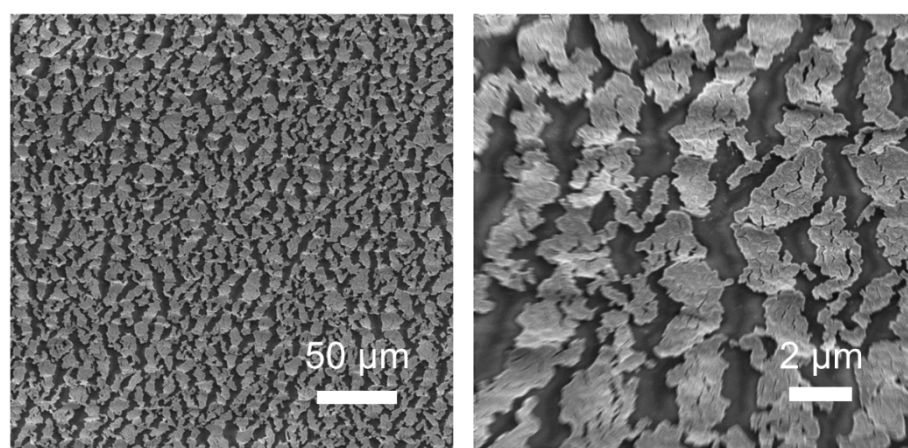


Supplementary Fig. 14: SEM images of the 80 nm-thick Au layer on the 1 μm -thick PDMS substrate without using the key the approach before and after post-annealing at 100 and 200 °C for 5 min. The SEM observation was repeated 3 times.

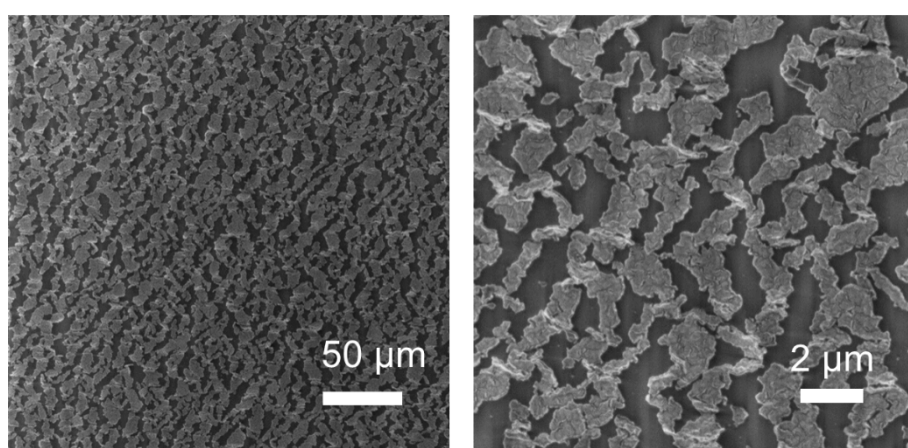
50% tensile strain



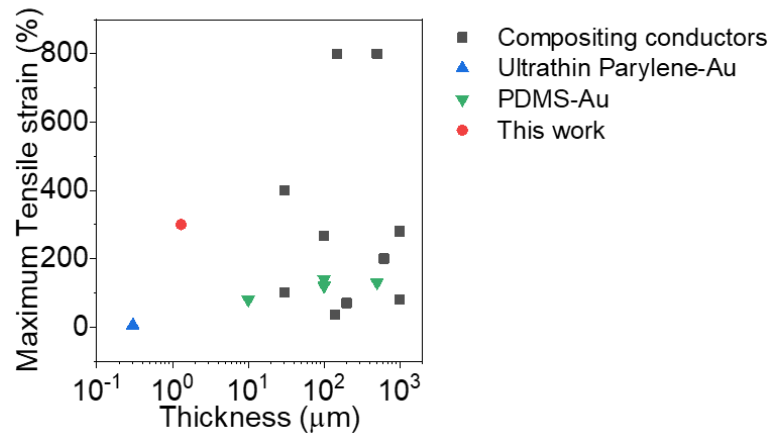
100% tensile strain



200% tensile strain

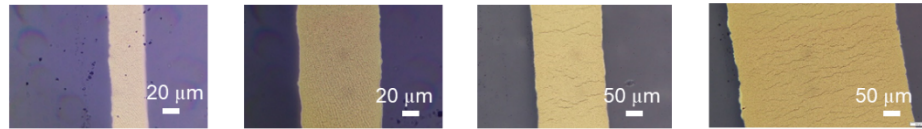


Supplementary Fig. 15: SEM images of the ultrathin PDMS-Au conductor under 50%, 100%, and 200% tensile strains. The SEM observation was repeated 3 times.

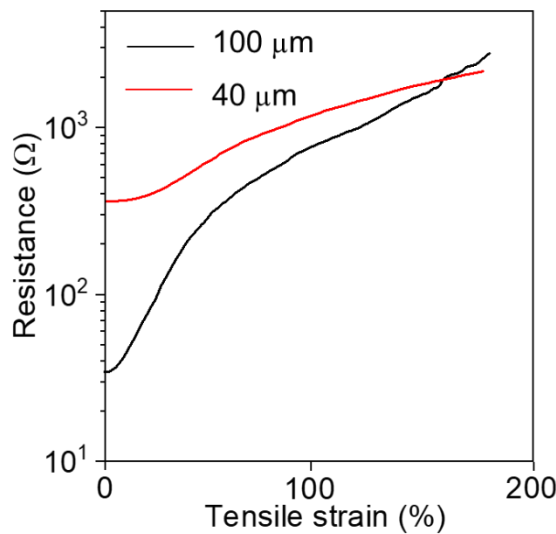


Supplementary Fig. 16: A comparison of the stretchability and thickness between the ultrathin PDMS-Au conductor and previous reports²⁻¹⁴. Note that the ultrathin PDMS-Au conductor shows a much higher stretchability than previously reported PDMS-Au with a thickness ranging from 10 to 500 μm. Other data points with larger stretchability belong to previously reported stretchable conductors with compositing components, including elastomer and conductive materials, such as metal nanowires, graphene, and carbon nanotubes. The ultrathin conductor with a very small stretchability is parylene-Au. Through the above comparison, we can conclude that achieving a large stretchability for thin stretchable conductors is challenging.

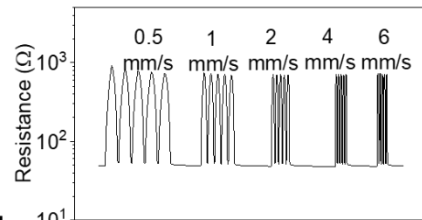
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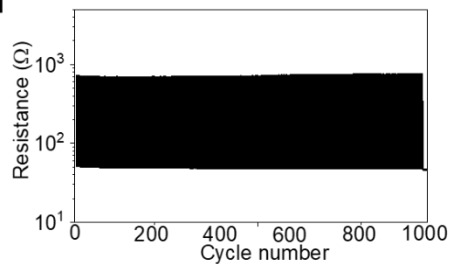
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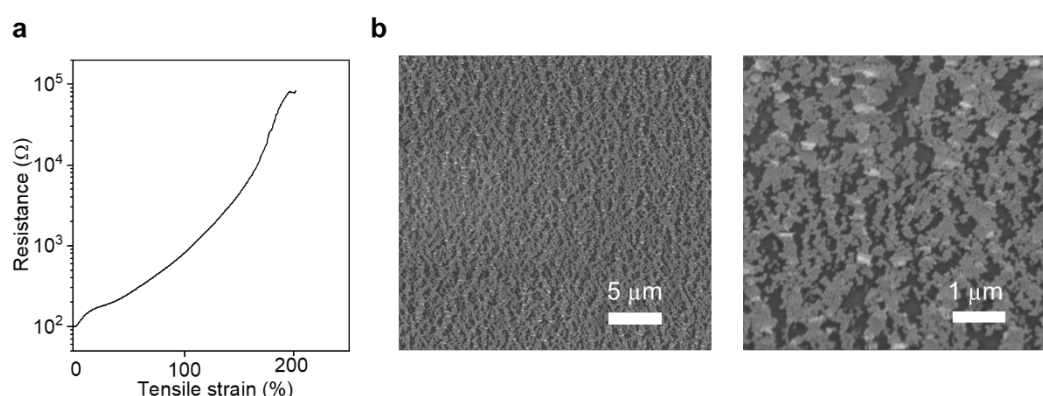
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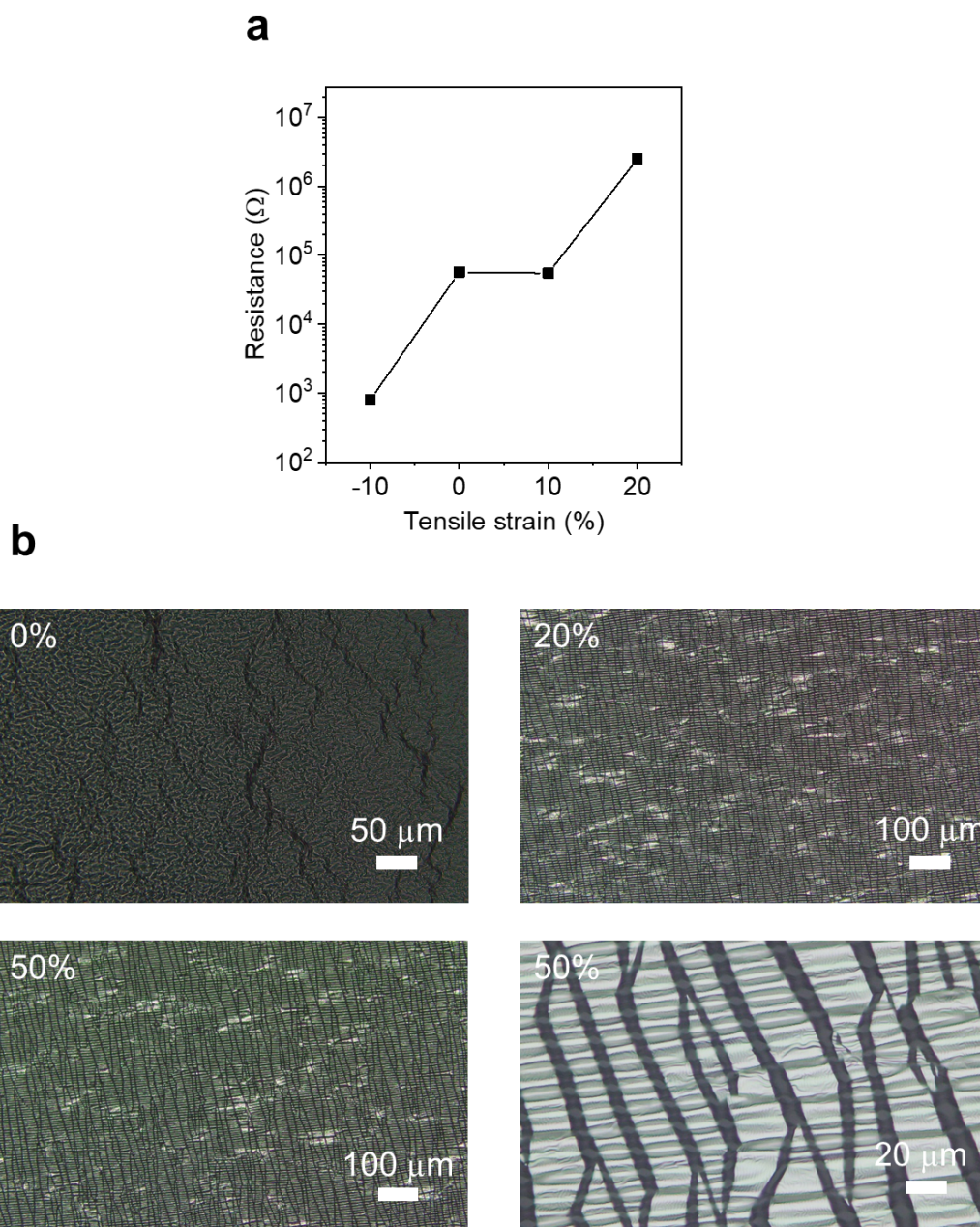
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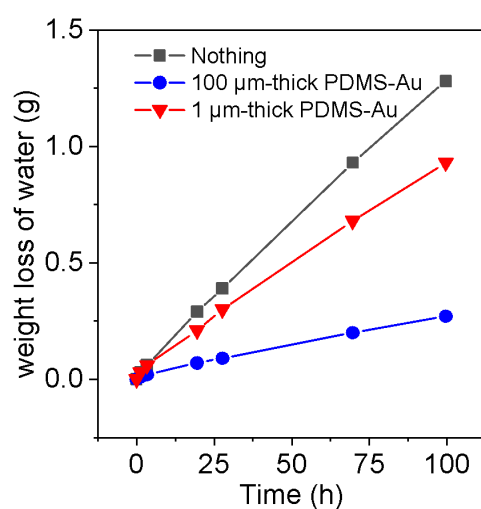
Supplementary Fig. 17: Patterning of the ultrathin PDMS-Au conductor using a shadow mask during the thermal evaporation process. **a**, Optical image of lines of a series of widths and a fixed length of 2 mm. **b**, A comparison of the electrical-mechanical property of the ultrathin PDMS-Au conductor of 40 and 100 μm width. **c, d**, Stability of the ultrathin PDMS-Au conductor with a line width of 100 μm (c) against a series of deformation speeds under 100% tensile strain, and (d) during 1000 cycles of the stretch/release process under 100% tensile strain.



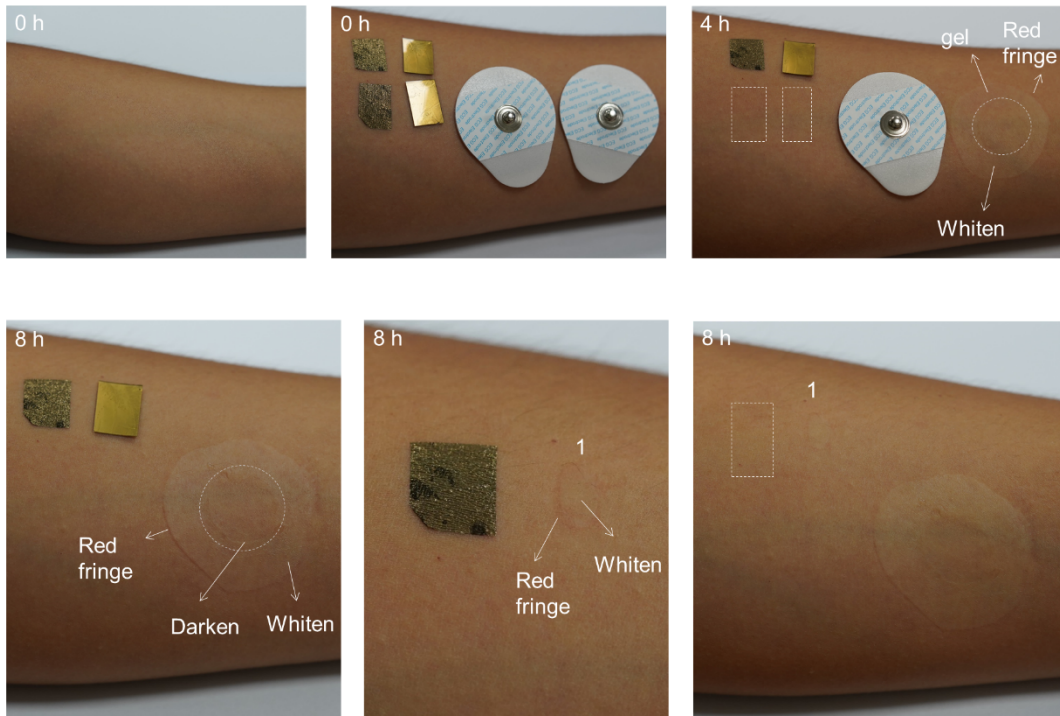
Supplementary Fig. 18: Characterisation of the micrometre-thick stretchable PDMS-Ag using the same fabrication method as the ultrathin PDMS-Au conductor. **a**, Electrical-mechanical property of the sample. **b**, SEM images of the PDMS-Ag under 100% tensile strain. The SEM observation was repeated 3 times.



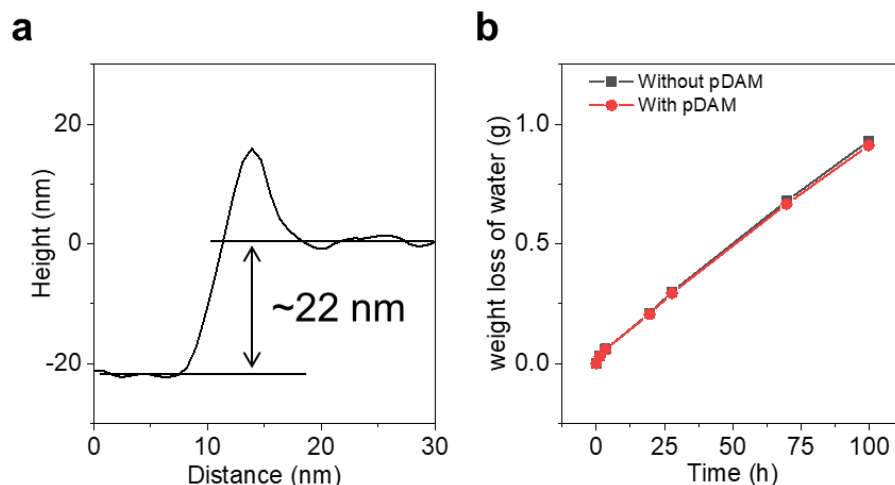
Supplementary Fig. 19: Characterization of the stretchable PDMS-ITO using the same fabrication method as the ultrathin PDMS-Au conductor. a, Electrical-mechanical property of the sample. The PDMS-ITO conductor showed a sharp resistance increase after being peeled off the supporting glass substrate and lost conductivity under 20% tensile strain. **b,** Optical images of the PDMS-ITO under a series of tensile strains. Clearly, there is only a random micrometre-scale buckling structure on the PDMS-ITO conductor after the fabrication. Moreover, the PDMS-ITO does not form micro-cracked morphology when stretched, leading to dramatically degraded conductivity under stretching. **The observation using optical microscopy was repeated 3 times.**



Supplementary Fig. 20: A comparison of the gas permeability between the ultrathin PDMS-Au electrode and thick electrode. The reference is a bottle that was uncovered, whereas the other two bottles were covered by the two films noted. The Au thickness for all samples was 50 nm. All the samples were kept in a stable environment of 37 °C and 55% relative humidity. Since the Au layer contains many cracks, it is breathable.

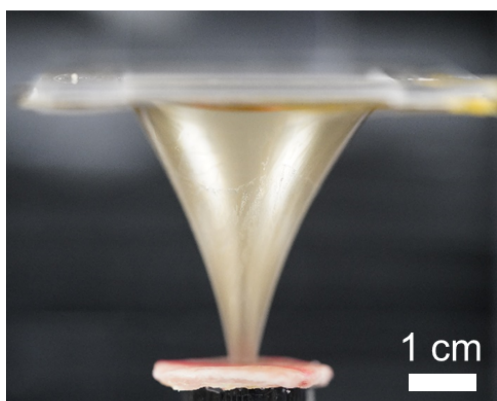


Supplementary Fig. 21: A comparison of skin before and after covered by the ultrathin PDMS-Au electrode, thick electrode, and commercial gel electrodes. After 4 h, skin covered by the commercial gel electrodes turned red at the gel region and the outer ring, while that covered by the ultrathin PDMS-Au conductor and 100 μm -thick PDMS-Au showed insignificant change. After 8 h, skin covered by the 100 μm -thick PDMS-Au turned red at the outer ring, while that covered by the ultrathin PDMS-Au conductor showed insignificant change. It is worthy to mention that skin covered by the commercial gel electrodes and 100 μm -thick PDMS-Au turned white because they were not breathable.

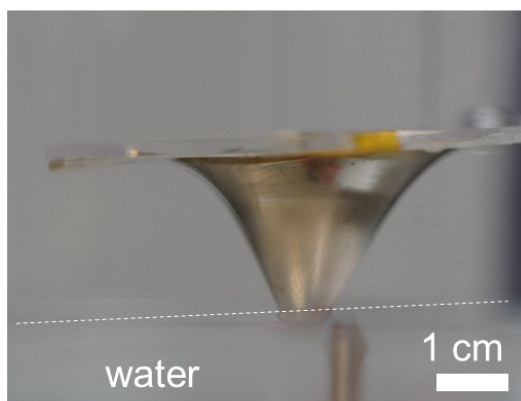


Supplementary Fig. 22: Characterization of the pDAM modified ultrathin PDMS-Au conductor. **a**, Surface profile result of the pDAM layer, showing a thickness of ~22 nm. **b**, A comparison of the gas permeability between the ultrathin PDMS-Au conductor and pDAM modified ultrathin PDMS-Au conductor, showing that the ultrathin pDAM does not influence the gas permeability at all.

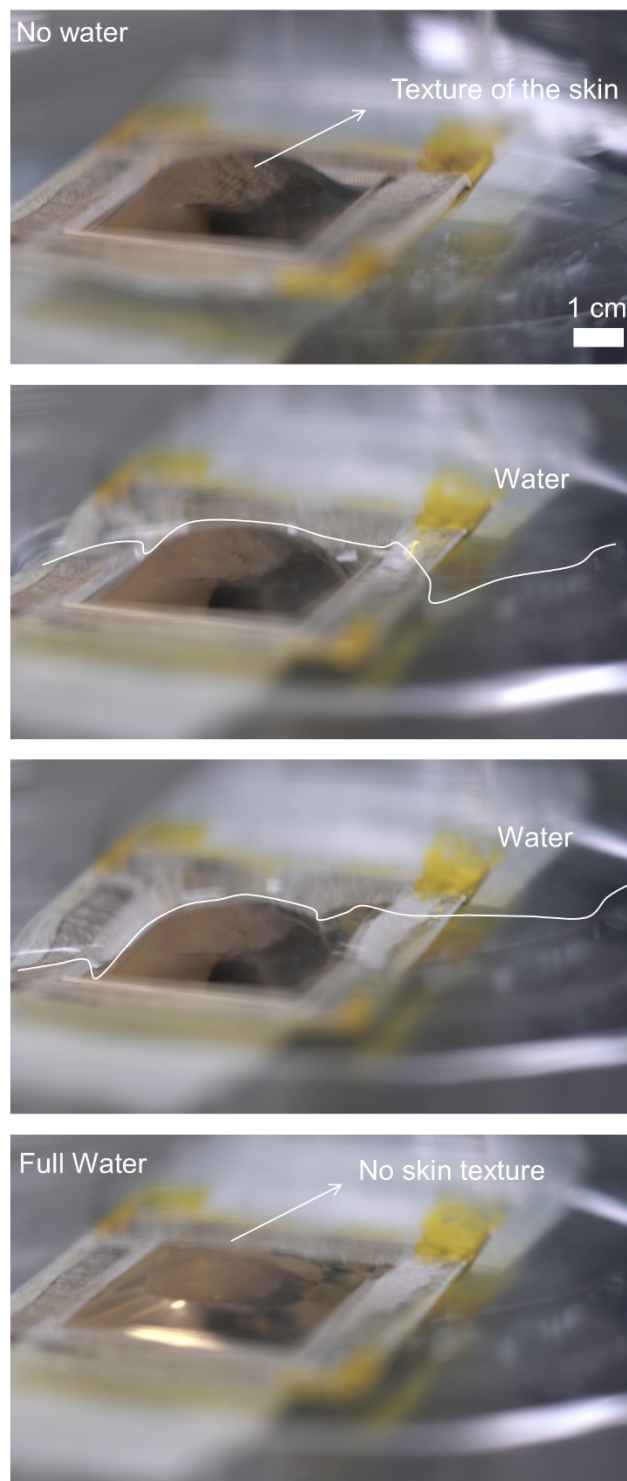
Ultrathin PDMS-Au-pDAM
Dry condition



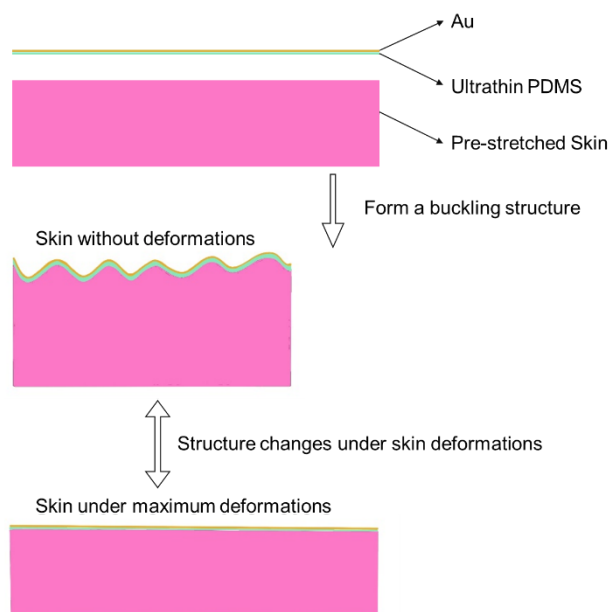
Ultrathin PDMS-Au-pDAM
In water after 30 min



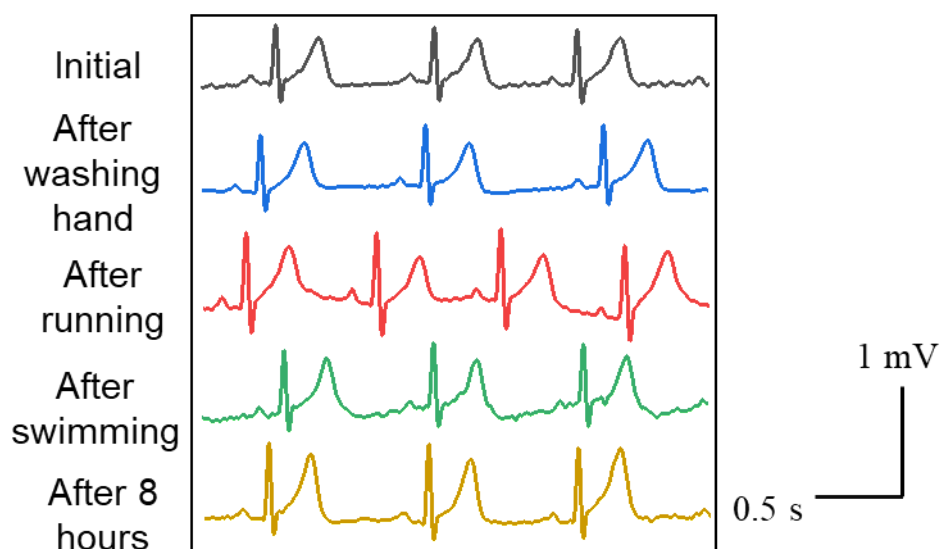
Supplementary Fig. 23: Optical images of ultrathin PDMS-Au-pDAM and skin during the peeling test.



Supplementary Fig. 24: Optical images of the ultrathin PDMS-Au conductor attached on the skin when gradually adding water. It is clear that there was skin texture on the ultrathin PDMS-Au conductor, indicating that a seamless contact was formed when there was no water. By gradually adding water, the skin texture on the ultrathin PDMS-Au conductor disappeared because water entered into the interface between the ultrathin PDMS-Au conductor and skin. After the ultrathin PDMS-Au conductor was immersed in water, the contact between the ultrathin PDMS-Au conductor and skin disappeared.



Supplementary Fig. 25: Illustration of forming a skin-mimic texture of the ultrathin PDMS-Au conductor on human skin.



Supplementary Fig. 26: Real-time ECG signals recorded by the ultrathin PDMS-Au-pDAM at different status of the volunteer, including after washing hands, running, swimming, and long-time wearing.

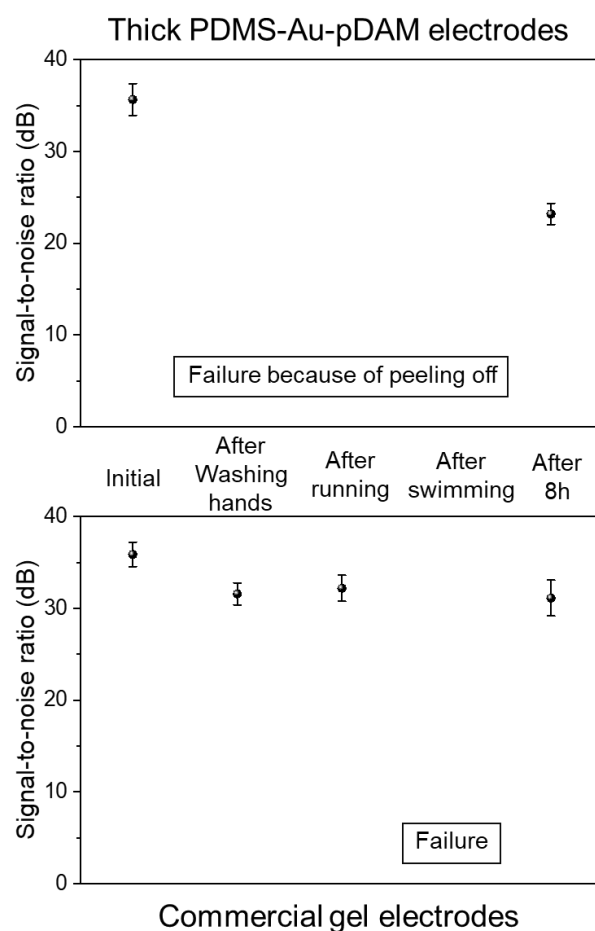
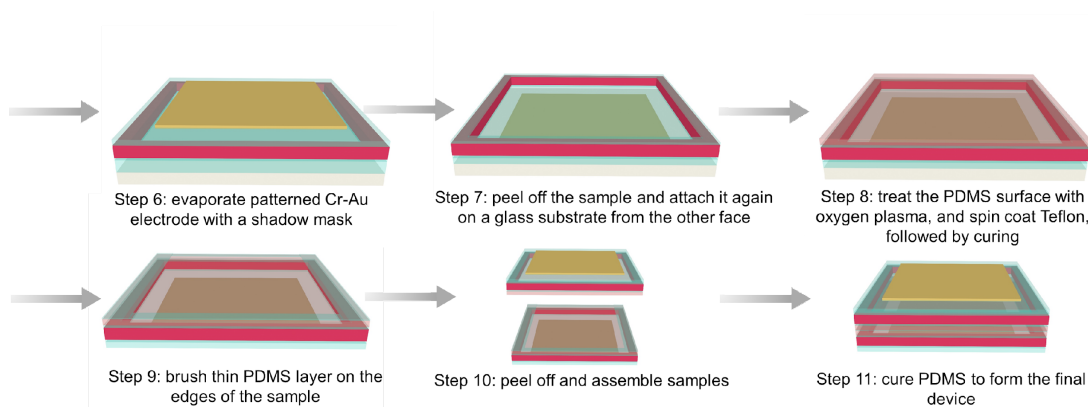


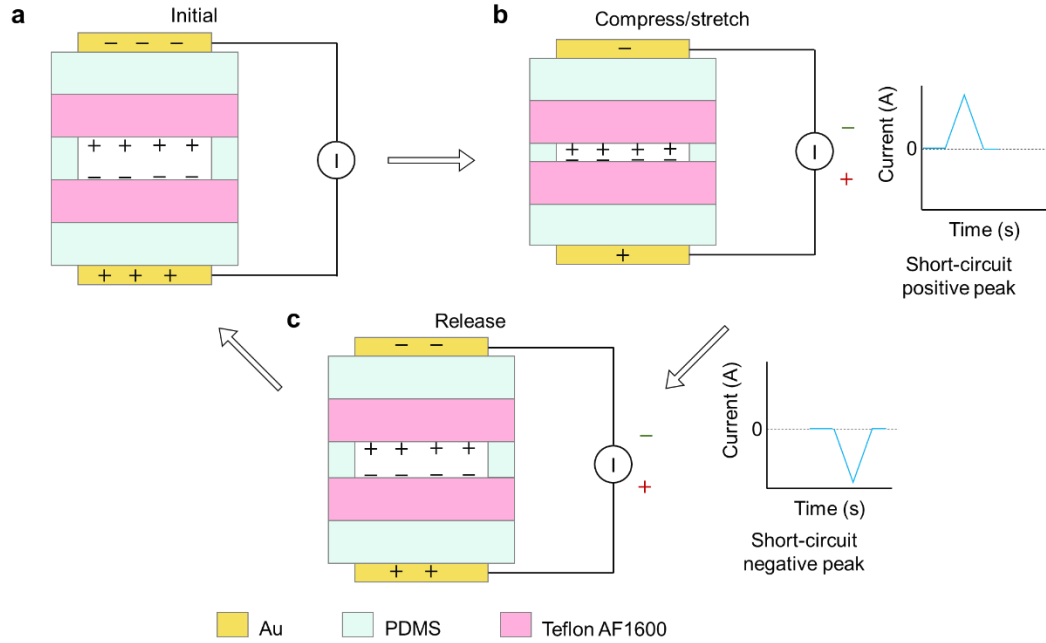
Figure S27. SNR values of ECG signals recorded by the thick PDMS-Au-pDAM electrodes and commercial gel electrodes at different statuses of the volunteer, including after washing hands, running, swimming, and long-time wearing ($n=3$ for all conditions). Data are presented as mean $\pm$ standard error of the mean (s.e.m.)



Supplementary Fig. 28: Optical image of 1 μm -thick PUNF-PDMS-Au-pDAM electrode attached onto an arm that was durable against heavy water washing.

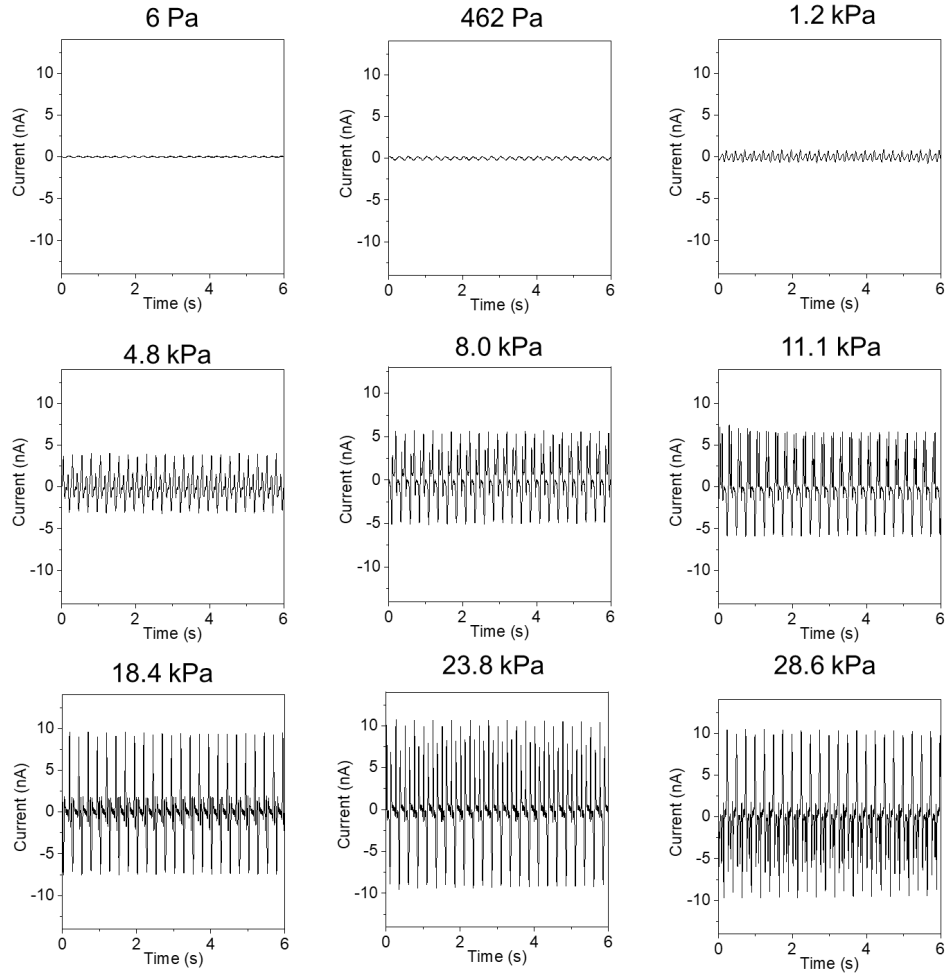


Supplementary Fig. 29: Fabrication process for the ultrathin stretchable self-powered sensor with the ultrathin PDMS-Au conductor, the fabrication process of which is described in Figure S4.

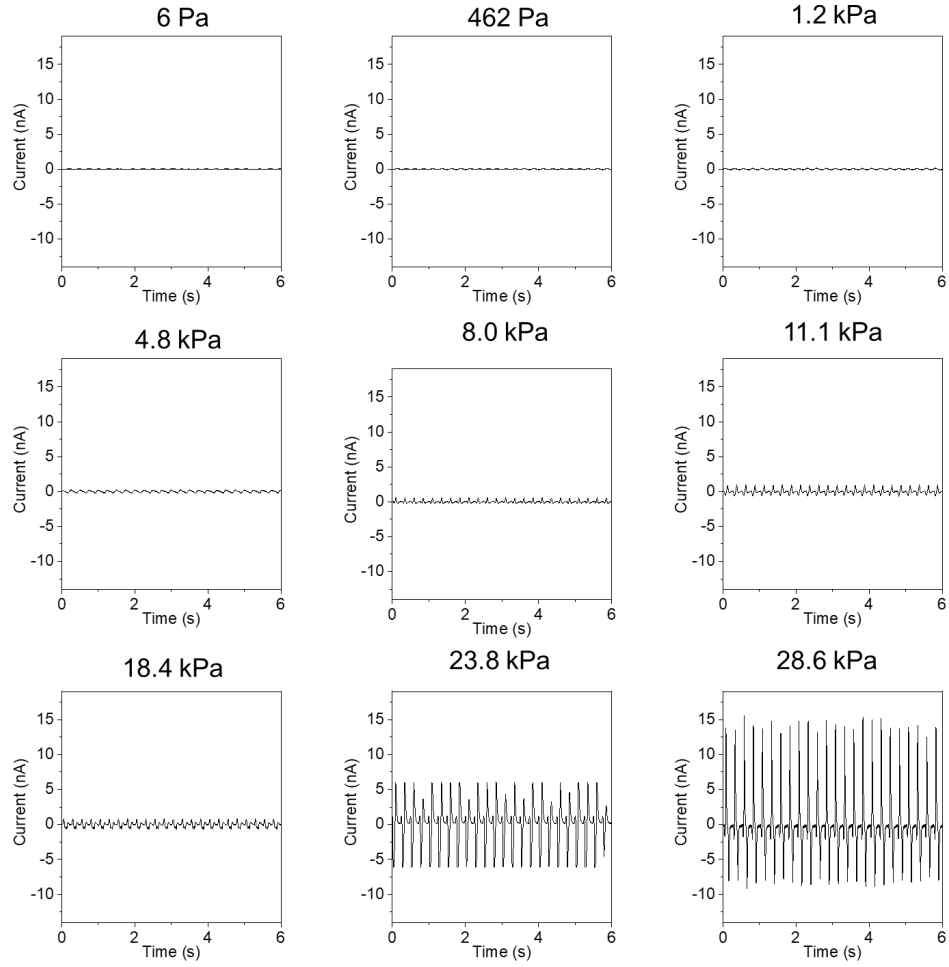


Supplementary Fig. 30. Schematic for the working mechanism of our ultrathin mechanical sensors. a. Initial state (no current output). b. Compressed by deformations, including pressure and stretch (the short-circuit current shows a positive peak). C. Recovery process by releasing the strain (the short-circuit current shows a negative peak).

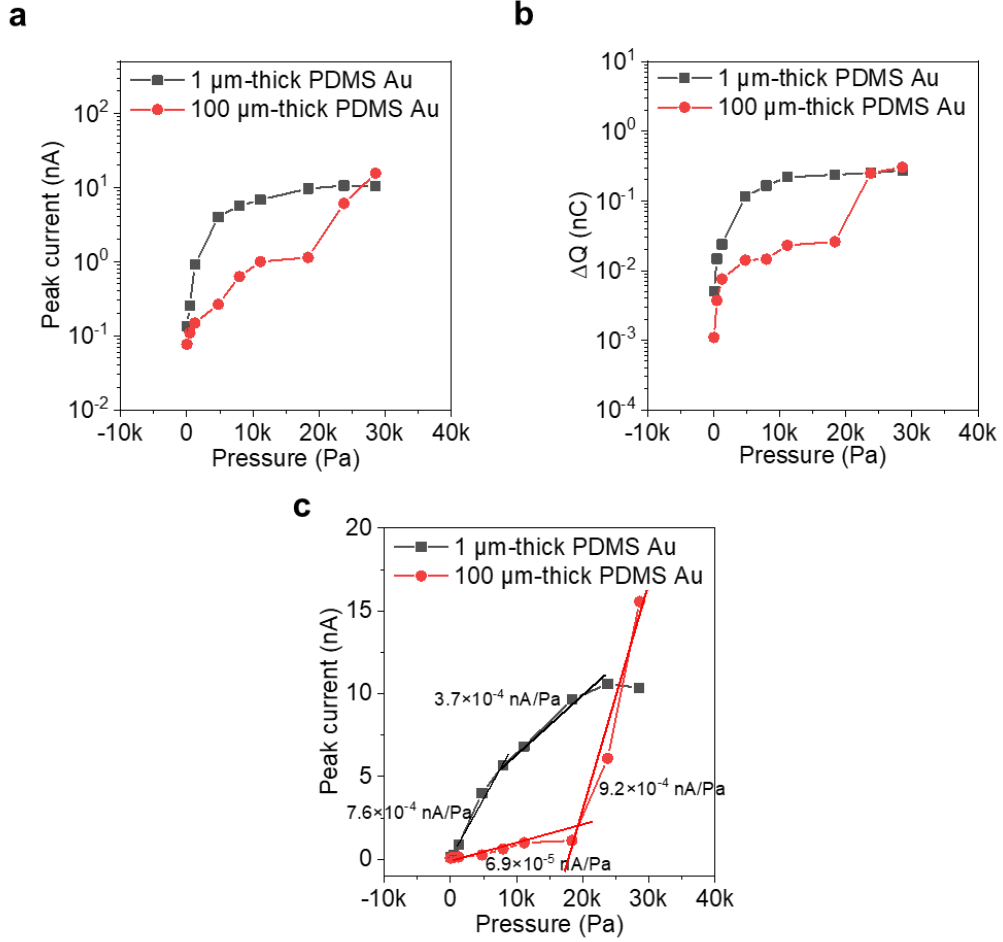
Mechanism description: The positive and negative electrostatic charges on the two Teflon AF1600 layers and the air gap compose the megascopic electric dipoles, which enable the sensor's with piezoelectric-like performance.^{15,16} In this structure, the dipole moment was determined by the thickness of the air gap. Initially there are induced charges ($\pm\sigma E$) on the two Au electrodes due to the electrostatic induction effect. When there is mechanical deformation caused by compression or stretch, the thickness of the air gap and dipole moment change, which induces free electrons to flow in the external load to generate alternating current (AC) outputs.



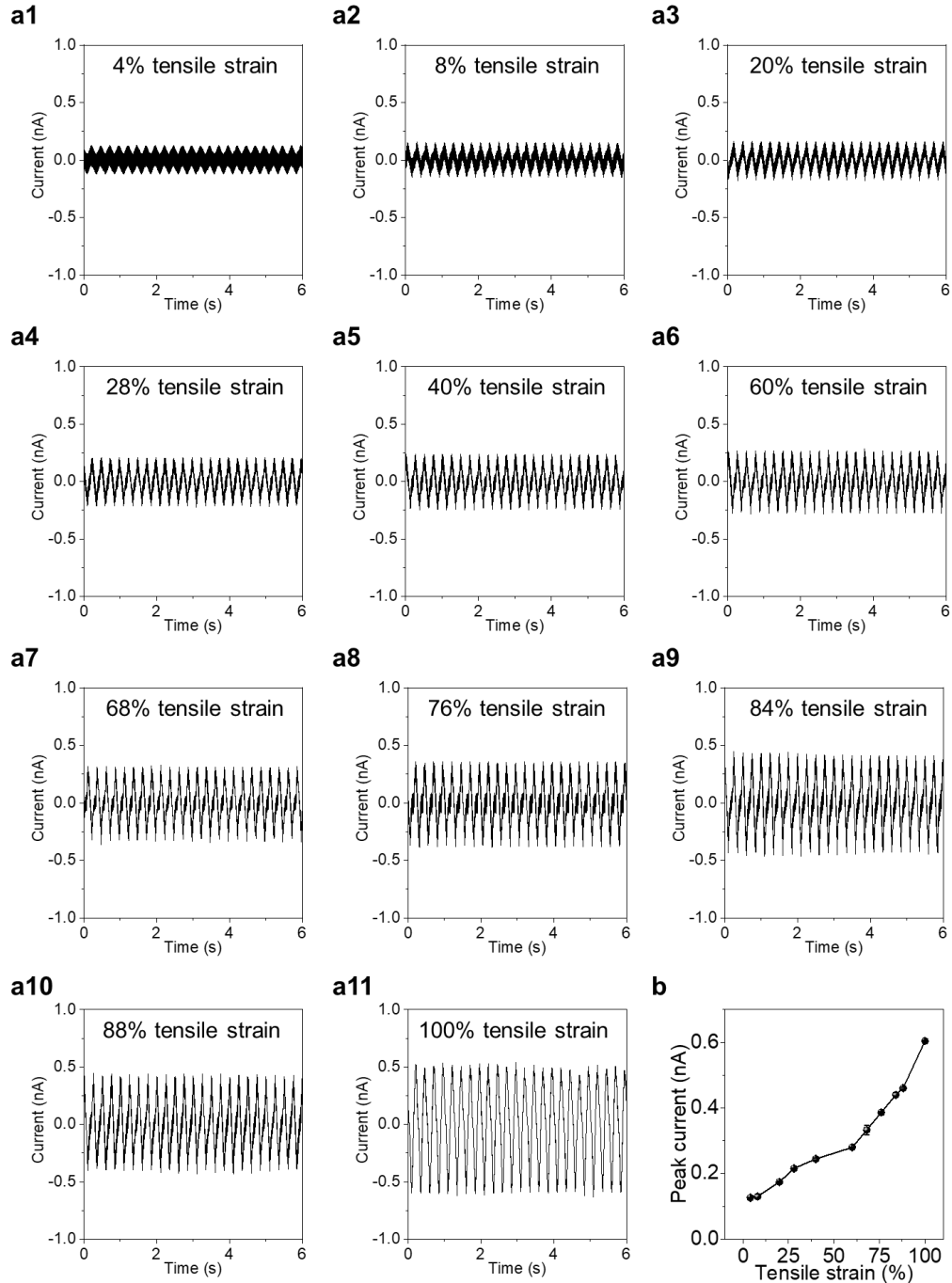
Supplementary Fig. 31: Output signals of the ultrathin stretchable self-powered pressure sensor with the ultrathin PDMS-Au conductor under a series of pressure values.



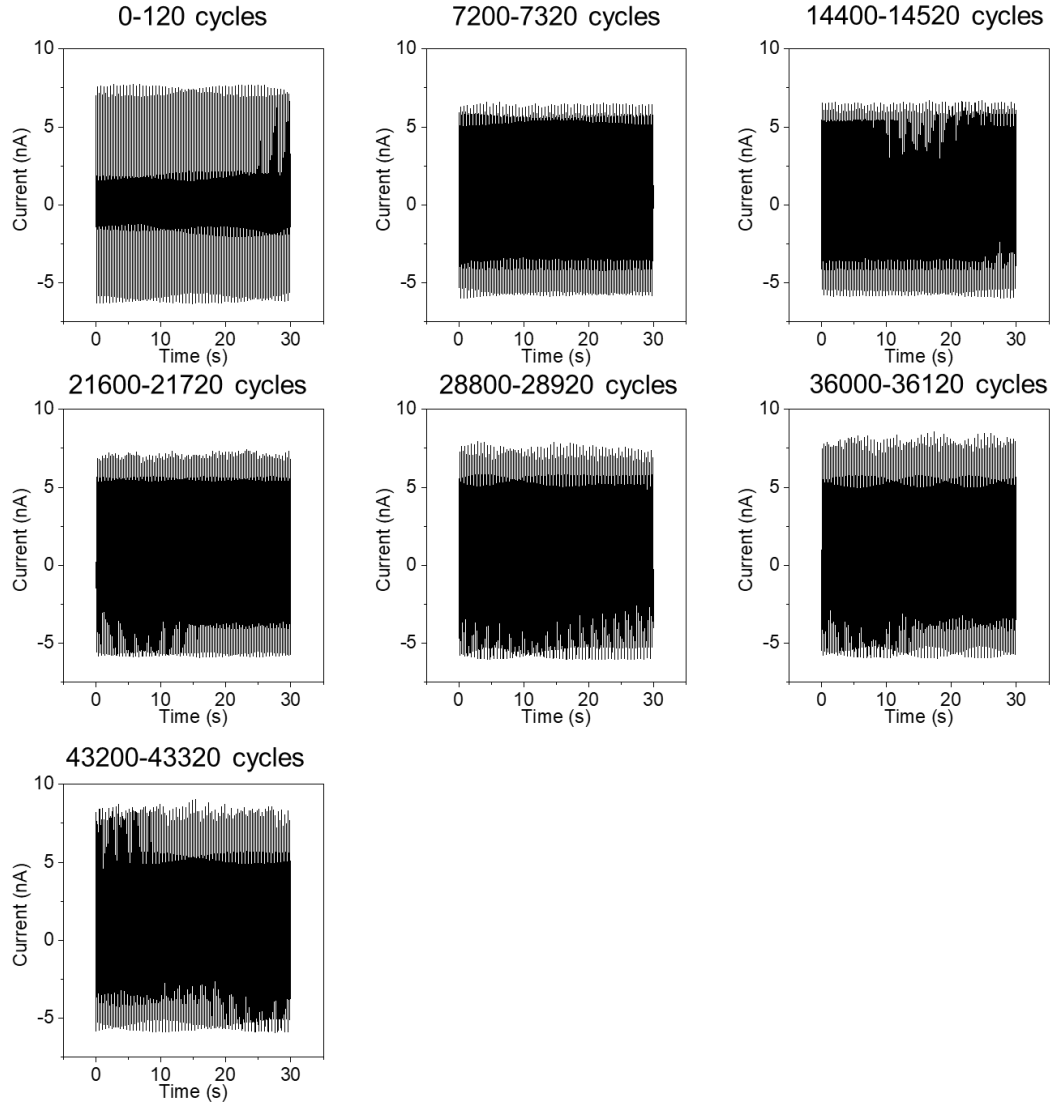
Supplementary Fig. 32: Output signals of the stretchable self-powered pressure sensor with the 100 μm -thick stretchable conductors under a series of pressure values.



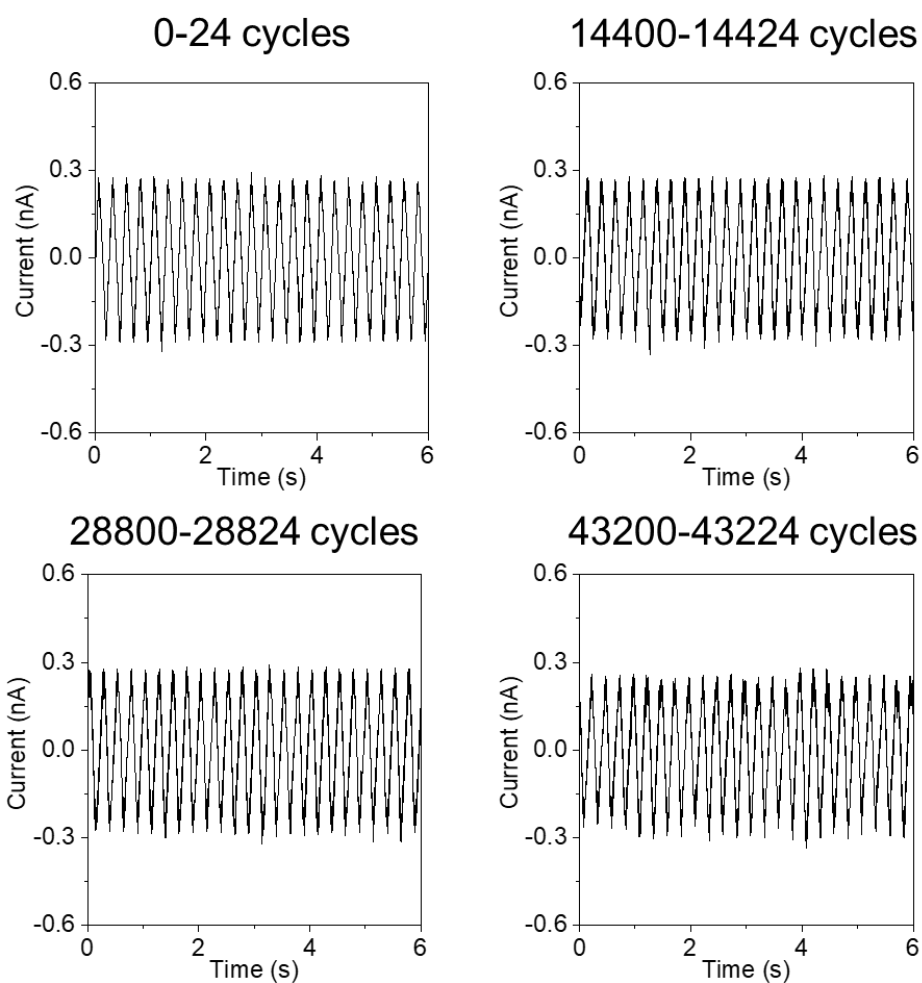
Supplementary Fig. 33: Performance comparisons of the stretchable self-powered sensor as a pressure sensor using the ultrathin PDMS-Au conductor and 100 μm -thick PDMS Au. a, Output peak current, and b, Simulated theoretical area transferred charges using finite element analysis. c, Sensitivity. Note the sensor area is same, i.e., 0.5 cm^2 .



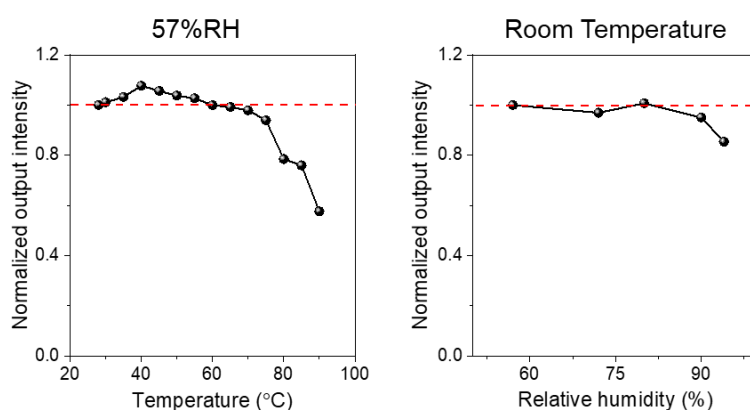
Supplementary Fig. 34: Characterization of the ultrathin stretchable self-powered sensors as a strain sensor. a1-a11, Output signals of the sensor under a series of tensile strains. b, Output current Vs tensile strain curve.



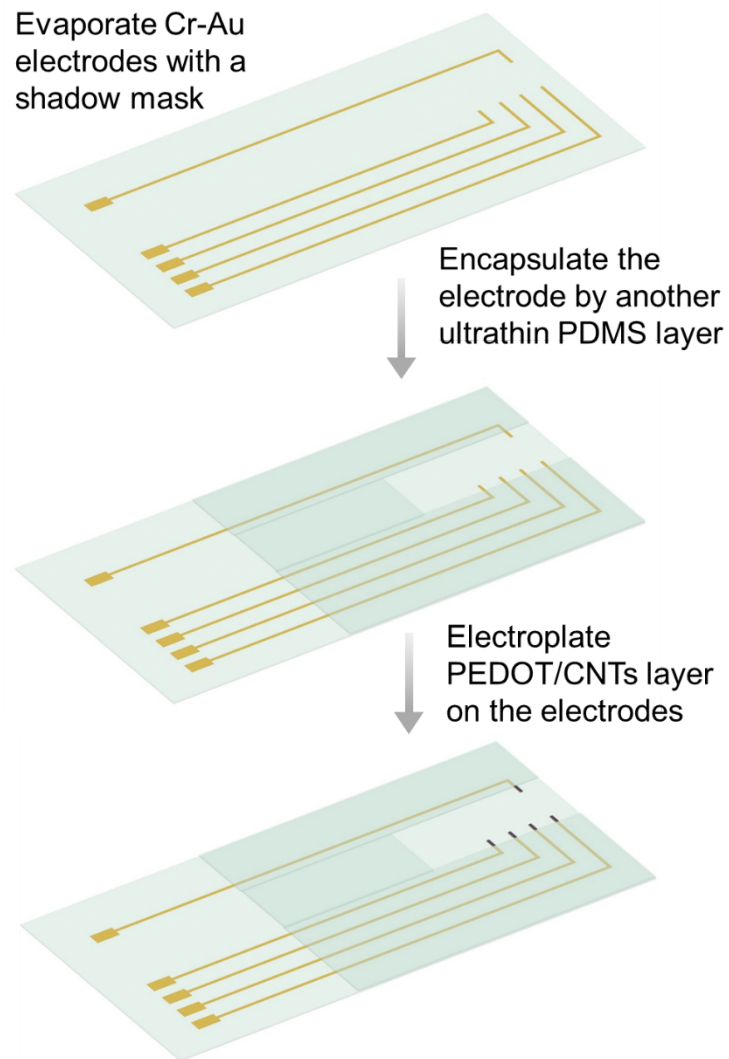
Supplementary Fig. 35: Output signals of the stretchable self-powered pressure sensor with the ultrathin PDMS-Au conductor during the cyclic test under a peak pressure of ~ 18.4 kPa.



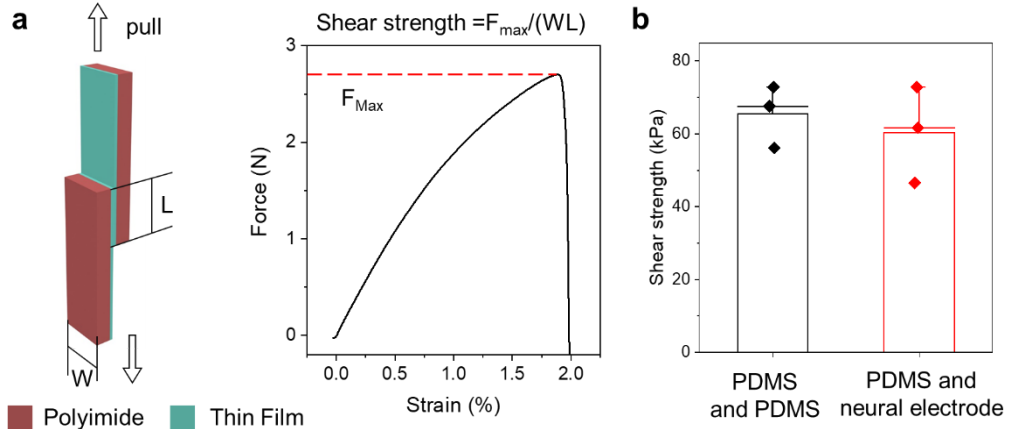
Supplementary Fig. 36: Output signals of the ultrathin stretchable self-powered strain sensor with the ultrathin PDMS-Au conductor during the cyclic test under a 60% tensile strain.



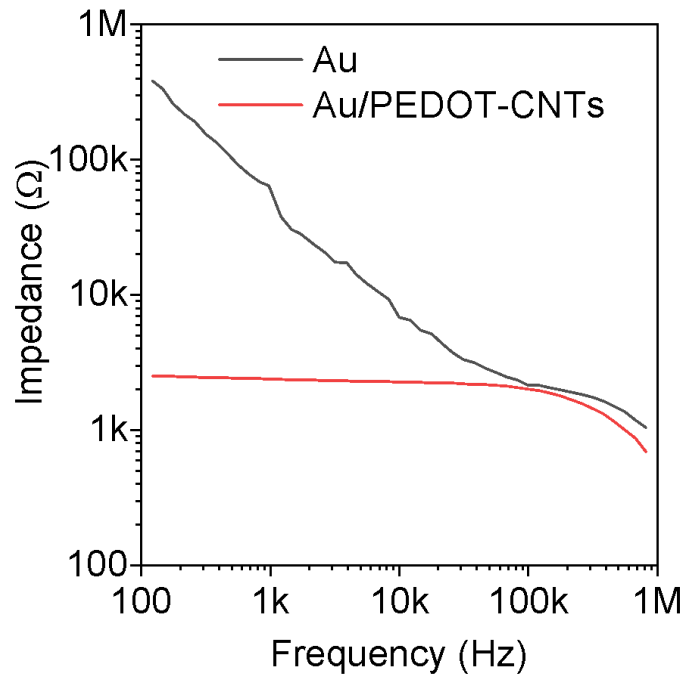
Supplementary Fig. 37: Environmental stability of the mechanical sensors under different temperatures and relative humidities.



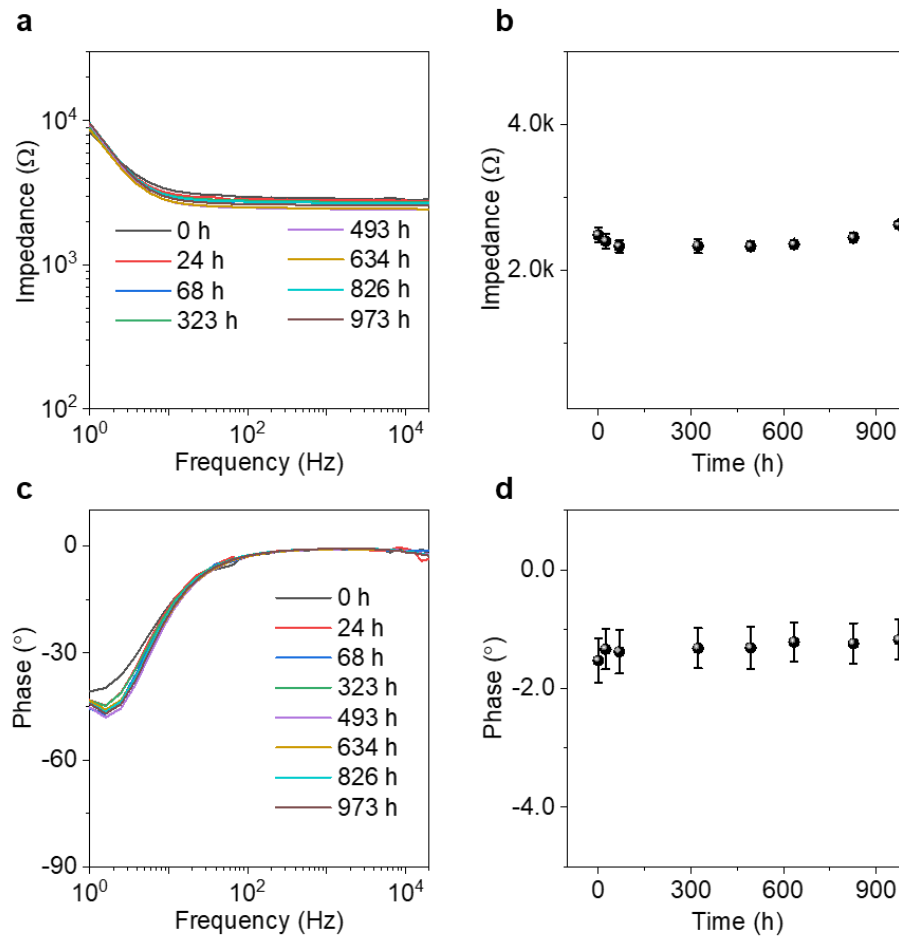
Supplementary Fig. 38: Illustration for the fabrication of the ultrathin PDMS-Au/PEDOT:CNTs neural electrodes.



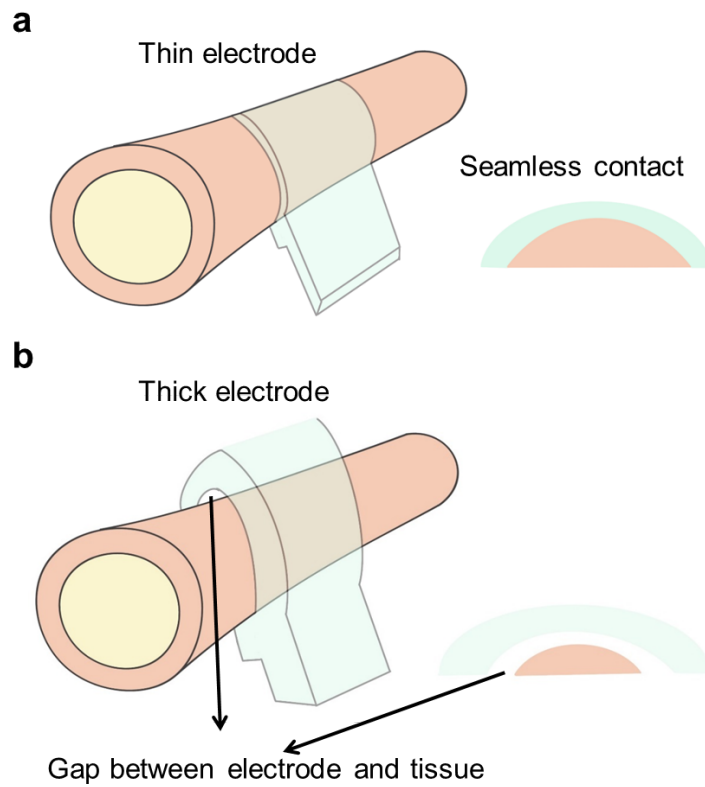
Supplementary Fig. 39: Characterization on the bonding strength of the ultrathin films. a. Setup illustration for measuring the shear strength. b. Shear strength of the bonded PDMS-PDMS samples and bonded and PDMS-PDMS-Au samples ($n=3$ for all conditions). Data are presented as mean $\pm$ standard error of the mean (s.e.m.)



Supplementary Fig. 40: Impedance measurement of the ultrathin PDMS-Au conductor/PEDOT:CNs neural electrodes, showing much lower impedance values than that of the ultrathin PDMS-Au conductor.

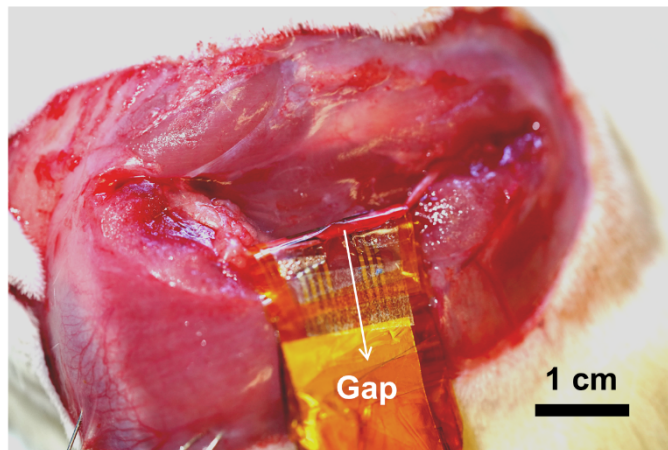


Supplementary Fig. 41: Long-term stability of the ultrathin neural electrodes immersing in saline at room temperature. A, C. EIS curves of the ultrathin neural electrodes measured at different time. B, D. Impedance and phase changes over time of the ultrathin neural electrodes at 1 kHz ($n=4$ for all conditions). Data are presented as mean $\pm$ standard error of the mean (s.e.m.)

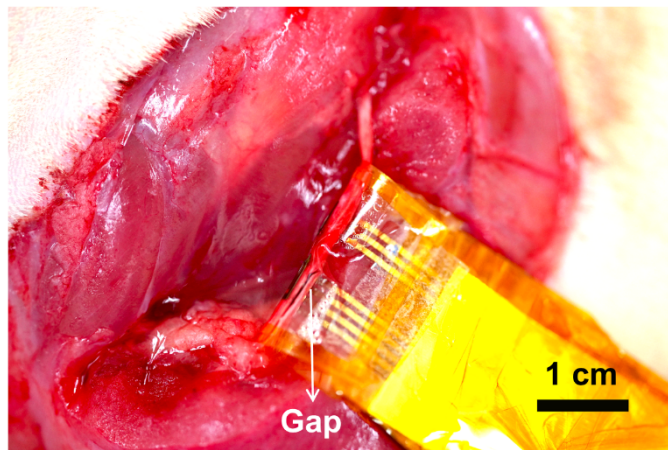


Supplementary Fig. 42: Illustration of the contact between electrodes and nerves.
a, Ultrathin electrode and **b**, Thick electrode.

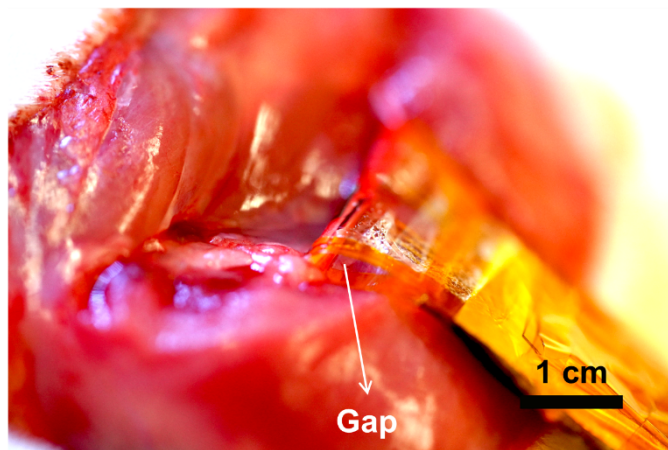
Front-top view



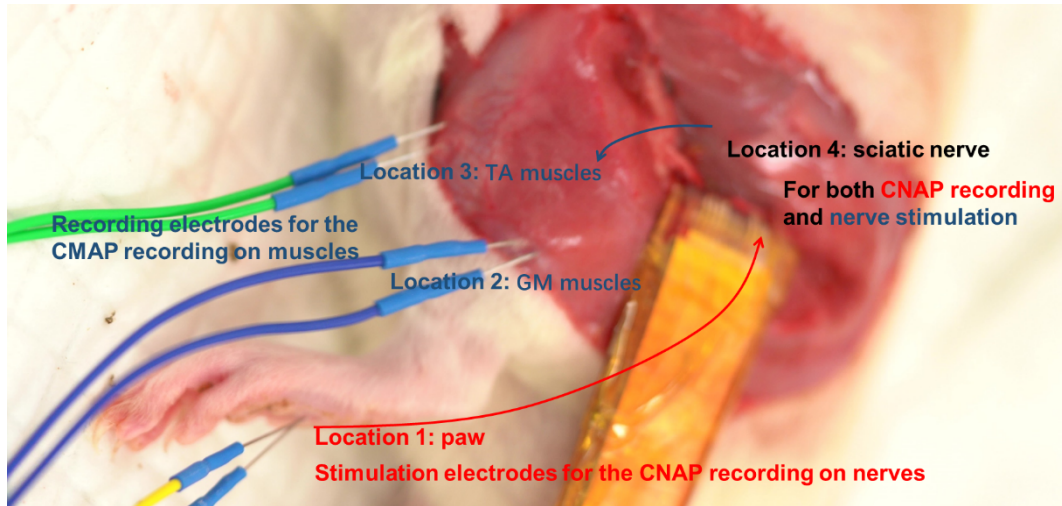
Left-top view



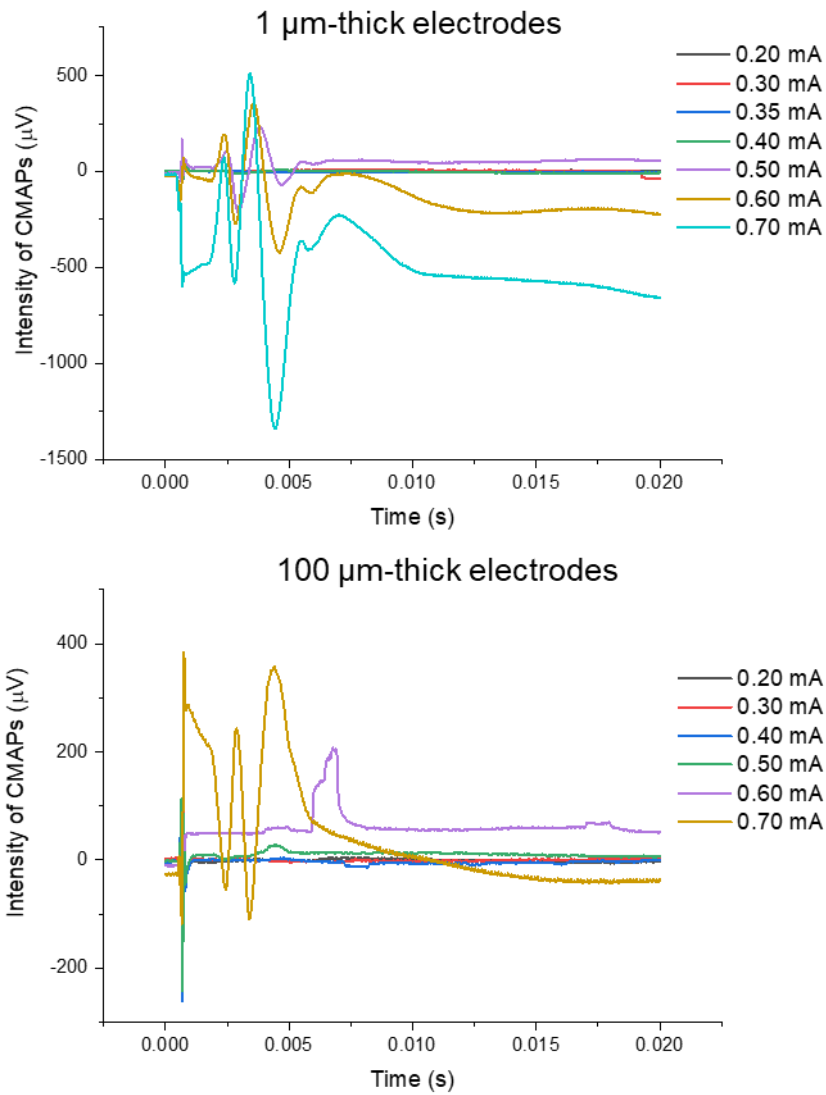
Left view



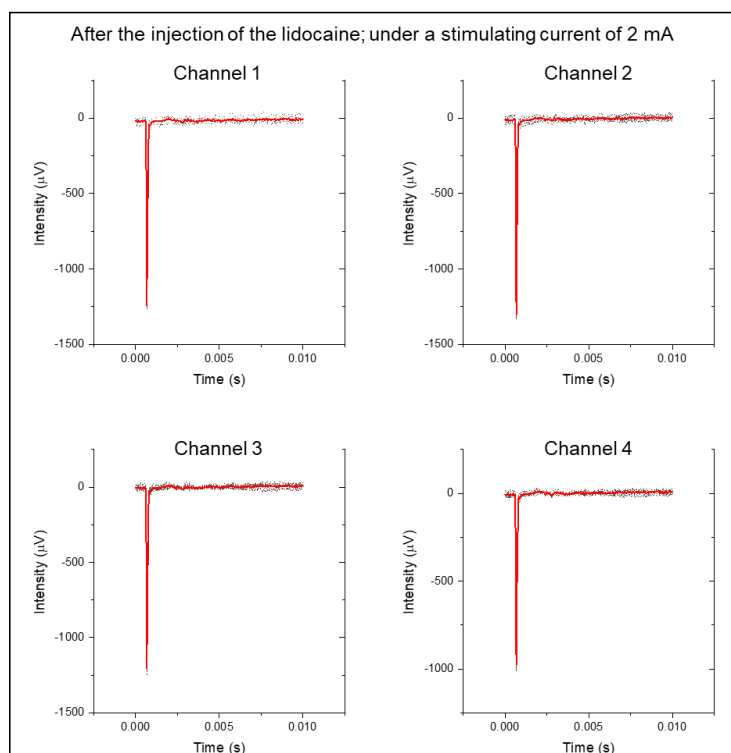
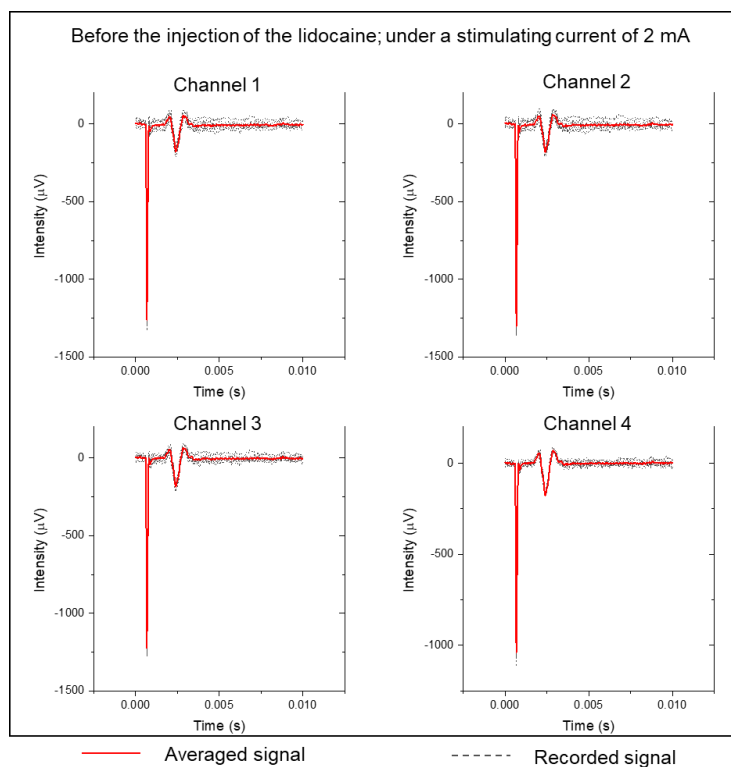
Supplementary Fig. 43: Optical images of the 100 μm-thick PDMS-Au-PEDOT:CNTs electrodes attached onto the nerve, showing gaps at the interface between themselves and the nerve tissues.



Supplementary Fig. 44: An optical image of the ultrathin neural electrode attached onto the nerve and three pairs of steel electrodes inserted into different locations of the rat. For the CNAP recording process, an electric stimulus was given to the paw (location 1) and the CNAP signals were recorded by the ultrathin neural electrode on the nerve (location 4); For the CMAP stimulation process, an electric stimulus was given to the nerve through the ultrathin neural electrode (location 4), and the CMAP signals were recorded by the steel probes on muscles (locations 2 and 3).



Supplementary Fig. 45: The real-time CMAPs recorded on the muscles by stimulating the nerves using the ultrathin neural electrodes and the 100 μm -thick electrodes when increasing the stimulation current from 0.20 to 0.70 mA. It is worthy to note that each curve at a single stimulation current was obtained by averaging 10 signals, and the details can be found in the signal processing part.

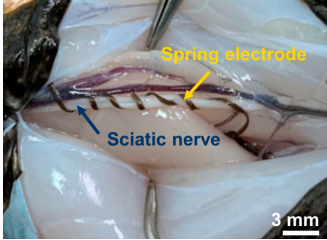
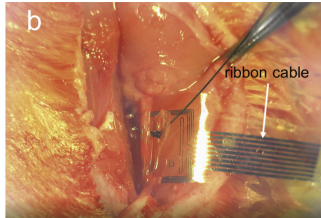


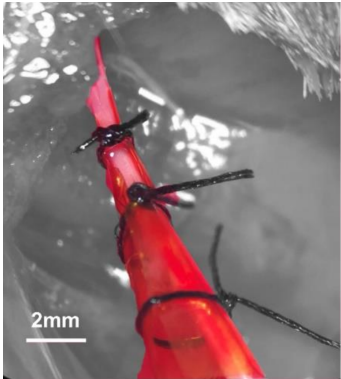
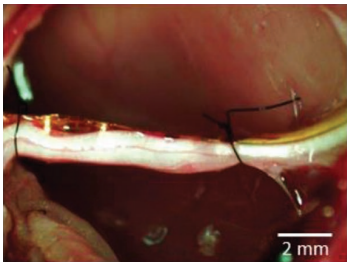
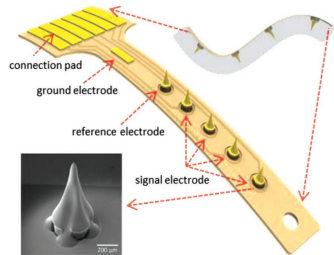
Supplementary Fig. 46: A comparison of CNAPs before and after the injection of the lidocaine for blocking the neural signals.

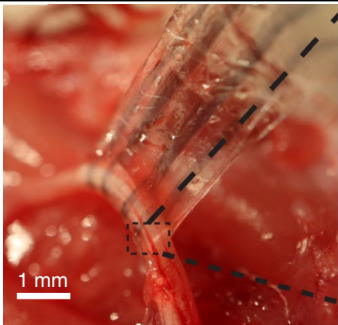
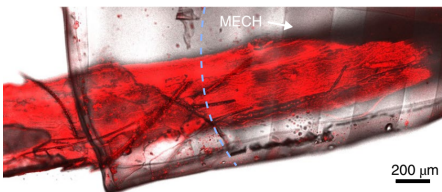
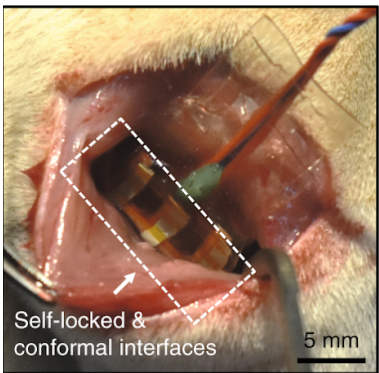
Supplementary Table 1: A comparison of the properties among previously reported conductors and the ultrathin PDMS-Au conductor.

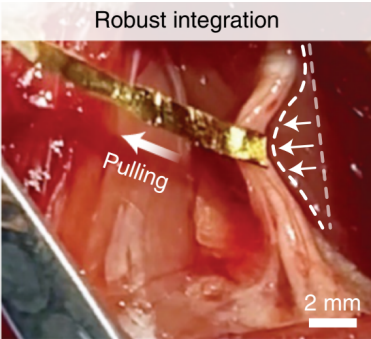
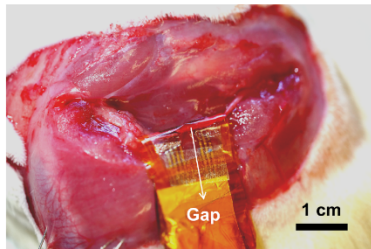
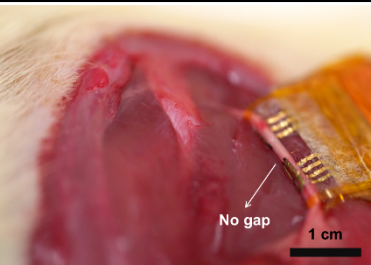
Materials	Substrate type	Thickness	Conductivity	Maximum strain	Durability at 0% strain	Resolution	Breathability	Scalability	Repeatability	Tensile force	Reference
Au nanomesh	—	100 nm	2×10^6 S/m	~30%	28% resistance increase after 500 cycles at 25% strain	500 μ m	Excellent	Bad	High	—	¹⁷
Au nanomesh/polyurethane nanofiber	Porous	1–3 μ m	~100 Ω /sq	~30%	5% resistance increase after 1000 cycles at 20% strain	1 mm	Excellent	Bad	-	12 mN at 30% strain	¹⁸
Au nanomesh/PDMS-polyurethane nanofiber	Porous	430 nm	~1.2 Ω /sq	~70%	3% resistance increase after 1000 cycles at 60% strain	2 cm	Excellent	Good	-	55 mN at 60% strain	¹⁹
Ag nanowire/polyurethane nanofiber	Porous	3 μ m	9190 S·cm ⁻¹	310%	82% resistance increase after 1000 cycles at 70% strain	—	Excellent	Bad	High	—	²⁰
Ag nanowire/SEBS	Porous	100 μ m	11000 S·cm ⁻¹	100%	~6% resistance increase after 2,000 cycles at 50% strain	1 mm	Good	Good	-	—	²¹
Au/parylene	Thin-film	300 nm	—	<5%	3% resistance increase after 1000 cycles of bending test		Bad	Bad	High	240 mN at 1% strain	³
Au/PDMS	Thin-film	100 μ m	11.8 Ω /sq	~140%	61% resistance increase after 1000 cycles at 100% strain	100 μ m	—	Good	High	1 N at 100% strain	⁴
Au/PDMS	Thin-film	10 μ m	—	~80%	—	—	—	Good	High	—	²²
Ag-Au nanowire/SBS	Thin-film	—	41850 S·cm ⁻¹	266%	110% resistance increase after 3000 cycles at 30% strain	—	Bad	Bad	-	—	⁵
Ag flakes/rubber	Thin film	30 μ m	4919 S·cm ⁻¹	400%	No conductivity after 500 cycles at 50% strain	500 μ m	Bad	Good	-	0.45 N at 100% strain	⁶
Au/PDMS	Thin-film	1.3 μ m	1.58×10^6 S/m (12.7 Ω /sq)	300%	1.7% resistance increase after 5000 cycles at 100% strain	40 μ m	Excellent	Good	High	38.9 mN at 100% strain	This work

Supplementary Table 2: A comparison of the ultrathin neural electrode with previously reported cuff-type flexible and stretchable nerve electrodes.

Material/Structure	Image	thickness	Plastic or elastic	Young's modulus	Contact quality	Selectivity	Fixing method	Function	Ref
GO-sodium alginate/spring		200 μm	Elastic	0.2-20 GPa	Obvious gap	None	Self	Recording	²³
Parylene-Au/self-locking		20 μm	Plastic	2-3 GPa	Obvious gap	4 Channels	Self	Recording	²⁴ ²⁵
SMP-PI-Au/serpentine		$\sim$ 50-200 μm	Plastic	Several GPa (PI); 400 kPa (SMP)	No gap	None	Tape	Recording	²⁶

PI-Au-PEDOT:CNT			~10-20 μm	Plastic	Several GPa	Obvious gap	4 Channels	Suture	Recording	²⁷
Polyimide strip-like structure with	 		20 μm	Plastic	2.5 GPa	Insert; No gap	5 Channels	Suture	Recording + Stimulation	²⁸

Self-healing and viscoplastic film/PEDOT:PSS		~120 μm	Viscoplastic	0.4-1.5 MPa	Obvious gap	None	Self	Stimulation	²⁹
Elastic polymer (PFPE-DMA)/PEDOT:PSS		No data	Elastic	~30 kPa	Obvious gap	None	Self	Stimulation	³⁰
PDMS-MPU0.4-IU0.6 Ag flake compositing electrode		~0.9 mm	Elastic	~170 kPa	Obvious gap	None	Self	Recording + Stimulation	³¹

PI-Au + electrical bioadhesive		50 μm for PI; No data for bioadhesive	Plastic	2-3 GPa	No gap	None	Self	Stimulation	³²
PDMS-Au		$\sim 100 \mu\text{m}$	Elastic	1-5 Mpa	Obvious gap	4 channels	Self	Recording + Stimulation	This work
PDMS-Au		$\sim 2.5 \mu\text{m}$	Elastic	1-5 Mpa	No gap	4 channels	Self	Recording + Stimulation	This work

Captions for the Supplementary Videos:

Supplementary Video 1: Laminating then ultrathin PDMS film onto the thick PDMS coated glass substrate. A seamless interface can happen automatically if the surfaces of the samples are clean. Speed: 5X.

Supplementary Video 2: Stretch test of the ultrathin conductor. The sample showed good recovery after a maximum tensile strain of 300%. Speed: 5X.

Supplementary Video 3: Adhesion test of the ultrathin electrode when the volunteer washed hands. With the modification of the pDAM layer, the electrode was very table. In contrast, without the modification, the electrode shrunk. Speed: 5X.

Supplementary Video 4: Adhesion test of the thick electrode when the volunteer washed hands. It was easily washed away by water. Speed: 5X.

Supplementary Video 5: Resistance measurement on the ultrathin electrode when moving the arm. Clearly, there was only little resistance change under fierce movement. Speed: 5X.

Supplementary Video 6: Demonstration of powering a liquid-crystal display (LCD) using the ultrathin mechanical sensor. It only needs a breath to deform the sensor to provide enough power for the LCD. Speed: 1X.

Supplementary Video 7: Selective muscle activation by stimulating the nerve using ultrathin neural electrodes. Speed: 5X.

Supplementary Video 8: Selective muscle activation by stimulating the nerve using thick electrodes. Speed: 5X.

Supporting References:

- 1 Ji, S. *et al.* Water-Resistant Conformal Hybrid Electrodes for Aquatic Endurable Electrocardiographic Monitoring. *Advanced materials* **32**, e2001496, doi:10.1002/adma.202001496 (2020).
- 2 Liu, Z. *et al.* Highly Stable and Stretchable Conductive Films through Thermal-Radiation-Assisted Metal Encapsulation. *Advanced materials* **31**, e1901360, doi:10.1002/adma.201901360 (2019).
- 3 Nawrocki, R. A. *et al.* Self-Adhesive and Ultra-Conformable, Sub-300 nm Dry Thin-Film Electrodes for Surface Monitoring of Biopotentials. *Advanced Functional Materials* **28**, 1803279, doi:10.1002/adfm.201803279 (2018).
- 4 Matsuhisa, N. *et al.* High-Transconductance Stretchable Transistors Achieved by Controlled Gold Microcrack Morphology. *Advanced Electronic Materials* **5**, 1900347, doi:10.1002/aelm.201900347 (2019).
- 5 Choi, S. *et al.* Highly conductive, stretchable and biocompatible Ag-Au core-sheath nanowire composite for wearable and implantable bioelectronics. *Nat Nanotechnol*, doi:10.1038/s41565-018-0226-8 (2018).
- 6 Matsuhisa, N. *et al.* Printable elastic conductors by in situ formation of silver nanoparticles from silver flakes. *Nature materials* **16**, 834-840, doi:10.1038/nmat4904 (2017).
- 7 Yue Wang, C. Z., Raphael Pfattner, Hongping Yan, Lihua Jin, Shucheng Chen, Francisco Molina-Lopez, Franziska Lissel, Jia Liu, Noelle I. Rabiah, Zheng Chen, Jong Won Chung, Christian Linder, Michael F. Toney, Boris Murmann, Zhenan Bao. A highly stretchable, transparent, and conductive polymer. *Science Advances* **3** (2017).
- 8 Yamada, T. *et al.* A stretchable carbon nanotube strain sensor for human-motion detection. *Nat Nanotechnol* **6**, 296-301, doi:10.1038/nnano.2011.36 (2011).
- 9 Chun, K. Y. *et al.* Highly conductive, printable and stretchable composite films of carbon nanotubes and silver. *Nature nanotechnology* **5**, 853-857, doi:10.1038/nnano.2010.232 (2010).
- 10 Sekitani, T. *et al.* Stretchable active-matrix organic light-emitting diode display using printable elastic conductors. *Nature materials* **8**, 494-499, doi:10.1038/nmat2459 (2009).
- 11 Xu, F. & Zhu, Y. Highly conductive and stretchable silver nanowire conductors. *Advanced materials* **24**, 5117-5122, doi:10.1002/adma.201201886 (2012).
- 12 Liang, J., Tong, K. & Pei, Q. A Water-Based Silver-Nanowire Screen-Print Ink for the Fabrication of Stretchable Conductors and Wearable Thin-Film Transistors. *Advanced materials* **28**, 5986-5996, doi:10.1002/adma.201600772 (2016).
- 13 Wang, Y. *et al.* Standing Enokitake-like Nanowire Films for Highly Stretchable Elastronics. *ACS Nano*, doi:10.1021/acsnano.8b05019 (2018).
- 14 Rujun Ma, B. K., Suik Cho, Minjun Choi, Seunghyun Baik. Extraordinarily High Conductivity of Stretchable Fibers of Polyurethane and Silver Nanoflowers. *ACS Nano* **9**, 10876-10886 (2015).
- 15 Zhong, J. *et al.* Smart Face Mask Based on an Ultrathin Pressure Sensor for Wireless Monitoring of Breath Conditions. *Advanced materials* **34**, e2107758, doi:10.1002/adma.202107758 (2022).
- 16 Chu, Y. *et al.* Human Pulse Diagnosis for Medical Assessments Using a Wearable Piezoelectret Sensing System. *Advanced Functional Materials* **28**,

-
- 1803413, doi:10.1002/adfm.201803413 (2018).
- 17 Miyamoto, A. *et al.* Inflammation-free, gas-permeable, lightweight, stretchable on-skin electronics with nanomeshes. *Nature nanotechnology* **12**, 907-913, doi:10.1038/nnano.2017.125 (2017).
- 18 Lee, S. *et al.* Ultrasoft electronics to monitor dynamically pulsing cardiomyocytes. *Nature nanotechnology* **14**, 156-160, doi:10.1038/s41565-018-0331-8 (2019).
- 19 Yan Wang, S. L., Tomoyuki Yokota, Haoyang Wang, Zhi Jiang, Jiabin Wang, Mari Koizumi, Takao Someya. A durable nanomesh on-skin strain gauge for natural skin motion monitoring with minimum mechanical constraints. *Sci. Adv.* **6**, eabb7043 (2020).
- 20 Jiang, Z. *et al.* Highly Stretchable Metallic Nanowire Networks Reinforced by the Underlying Randomly Distributed Elastic Polymer Nanofibers via Interfacial Adhesion Improvement. *Advanced materials* **31**, e1903446, doi:10.1002/adma.201903446 (2019).
- 21 Yadong Xu, B. S., Yun Ling, Qihui Fei, Zanyu Chen, Xiaopeng Li, Peijun Guo, Nari Jeon, & Shivam Goswami, Y. L., Shinghua Ding, Qingsong Yu, Jian Lin, Guoliang Huang, and Zheng Yan. Multiscale porous elastomer substrates for multifunctional on-skin electronics with passive-cooling capabilities. *Proc. Natl. Acad. Sci. U.S.A* **117**, 205-213, doi:10.1073/pnas.1917762116/-/DCSupplemental (2020).
- 22 Zhang, B. *et al.* Stretchable Conductive Fibers Based on a Cracking Control Strategy for Wearable Electronics. *Advanced Functional Materials* **28**, 1801683, doi:10.1002/adfm.201801683 (2018).
- 23 Zhang, M. Microribbons composed of directionally self-assembled nanoflakes as highly stretchable ionic neural electrodes. *PNAS*, doi:10.1073/pnas.2003079117/-/DCSupplemental (2020).
- 24 Cobo, A. M. Parylene-Based Cuff Electrode With Integrated Microfluidics for Peripheral Nerve Recording, Stimulation, and Drug Delivery. *JOURNAL OF MICROELECTROMECHANICAL SYSTEMS* (2019).
- 25 Elyahoodayan, S., Larson, C., Cobo, A. M., Meng, E. & Song, D. Acute in vivo testing of a polymer cuff electrode with integrated microfluidic channels for stimulation, recording, and drug delivery on rat sciatic nerve. *Journal of neuroscience methods* **336**, 108634, doi:10.1016/j.jneumeth.2020.108634 (2020).
- 26 ZHANG, Y. Climbing-inspired twining electrodes using shape memory for peripheral nerve stimulation and recording. *Science Advances* (2019).
- 27 Chen, N. *et al.* Nanotunnels within Poly(3,4-ethylenedioxythiophene)-Carbon Nanotube Composite for Highly Sensitive Neural Interfacing. *ACS nano* **14**, 8059-8073, doi:10.1021/acsnano.0c00672 (2020).
- 28 Wang, J. *et al.* A Highly Selective 3D Spiked Ultraflexible Neural (SUN) Interface for Decoding Peripheral Nerve Sensory Information. *Advanced healthcare materials* **7**, doi:10.1002/adhm.201700987 (2018).
- 29 Liu, Y. *et al.* Morphing electronics enable neuromodulation in growing tissue. *Nature biotechnology* **38**, 1031-1036, doi:10.1038/s41587-020-0495-2 (2020).
- 30 Liu, Y. *et al.* Soft and elastic hydrogel-based microelectronics for localized low-voltage neuromodulation. *Nature biomedical engineering* **3**, 58-68, doi:10.1038/s41551-018-0335-6 (2019).
- 31 Song, K. I. *et al.* Adaptive self-healing electronic epineurium for chronic bidirectional neural interfaces. *Nature communications* **11**, 4195,

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- doi:10.1038/s41467-020-18025-3 (2020).
- 32 Deng, J. *et al.* Electrical bioadhesive interface for bioelectronics. *Nature materials* **20**, 229-236, doi:10.1038/s41563-020-00814-2 (2021).