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Conformal photoplethysmography sensor with ultrathin architecture for accurate health monitoring

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Wearable photoplethysmography (PPG) technology is a straightforward method for comfortable and continuous health monitoring. However, the mechanical mismatch and optical noise at the skin–sensor interface compromise the accurate pulse waveform acquisition of PPG sensors. In this work, we propose a conformable PPG sensor to eliminate interfacial optical crosstalk for accurate pulse monitoring. The device possesses an ultrathin thickness of 5.2 μm and a light weight of 8.7 g m⁻², which enable conformal contact between the device and the skin. Due to the elimination of the light reflection noise through close contact and the high sensitivity of the perovskite photodetector, the PPG sensor achieves high pulse signal quality with a perfusion index of 4.4% and a low working illumination intensity of 25.3 $\mu\text{W cm}^{-2}$. Furthermore, heart rate variability analysis and continuous blood pressure monitoring are performed using these accurate pulse signals. This work provides a reliable interface engineering strategy for eliminating optical noise in wearable health sensing electronics.

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Introduction

Wearable healthcare devices can collect various physiological data of the human body, enabling early detection and timely

treatment of diseases.^{1–4} Heart rate (HR) and blood pressure are critical signals for monitoring and diagnosing health conditions, such as cardiovascular diseases.^{5–8} Photoplethysmography (PPG) technology is a non-invasive pulse detection method, which has been widely employed for the long-term monitoring of cardiovascular health conditions.^{9–11}

Current wearable PPG sensors commonly use the reflection mode to enable single-sided sensing at various body locations.¹² These sensors are integrated on rigid or flexible substrates with large thickness and bending stiffness.^{13–17} This causes a mechanical and optical mismatch at the interface between the skin and the sensor.^{18–20} Specifically, these PPG sensors cannot form conformal contact with the roughness of the epidermis without continuously applying large pressure due to their intrinsic stiffness. This will generate air gaps, which induce surface reflection of the incident light and the motion artifacts, eventually resulting in poor signal accuracy that hinders further hemodynamic analysis.^{21,22} Therefore, it is necessary for the device to adapt to the dynamic changes of the skin. Previous research has focused on the integration of optical microstructures and the use of high-power illumination to improve signal accuracy.^{23,24} A straightforward method is to minimize the thickness to reduce the device stiffness, enabling conformal contact with skin.

In this work, we propose a conformable PPG sensor to eliminate the interfacial optical crosstalk for highly accurate pulse monitoring. The device integrates a perovskite photodetector for weak light detection and an LED, with a total thickness of 5.2 μm

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and a weight of 8.7 g m^{-2} . These ultrathin and lightweight features enable the device to form conformal contact with the skin texture, avoiding light reflection noise caused by the air gaps. Combined with serpentine interconnects to ensure mechanical stability under various deformations, the device can achieve a high perfusion index of 4.4% and capture detailed pulse waveform features under a low working illumination intensity of $25.3 \text{ } \mu\text{W cm}^{-2}$. Furthermore, using the high-fidelity signal, heart rate variability analysis and blood pressure monitoring could be performed, confirming the capability of the device for cardiovascular health assessment.

Experimental

Materials

PEAI ($\geq 99.5\%$), MAI ($\geq 99.5\%$), and PbI_2 ($> 99.99\%$) were purchased from Xi'an Polymer Light Technology Corp., without further processing. *N,N*-Dimethylformamide (DMF, 99.9%) was purchased from Shanghai Aladdin Biochemical Technology Co., Ltd. All target materials for magnetron sputtering (Au and SiO_2) were purchased from ZhongNuo Advanced Material (Beijing) Technology Co., Limited.

Fabrication of the PPG sensor

First, the heavily-doped Si wafer was sonicated for 5 minutes in acetone and deionized water, respectively. Second, a $2.5 \text{ } \mu\text{m}$ parylene-C film was deposited on the Si wafer using a parylene-C vacuum deposition system (MAGGIE MQP-3001). Then, the patterned Au electrodes were prepared through lithography and magnetron sputtering (PVD75, Kurt J. Lesker). Subsequently, the patterned SiO_2 was obtained using the method given as Supplementary Note 1 in the SI. The perovskite precursor was prepared by combining 190.8 mg of PEAi, 199.2 mg of MAI, and 737.6 mg of PbI_2 (mole ratio: 3 : 2 : 4) in 1 ml of DMF, under stirring using a magnetic stirrer at a temperature of $70 \text{ } ^\circ\text{C}$ for 6 hours in an N_2 -filled glovebox. $100 \text{ } \mu\text{L}$ of perovskite precursor was spin-coated on the as-prepared substrate at 5000 rpm for 40 s, then immediately placed on a hot plate at $100 \text{ } ^\circ\text{C}$ for 10 min to remove residual solvents. Next, an LED was integrated on the designated electrodes. Finally, the prepared device was encapsulated by the $2.5 \text{ } \mu\text{m}$ parylene-C film using a parylene-C vacuum deposition system (MAGGIE MQP-3001).

Material characterization and device performance test

Scanning electron microscope images were obtained using a scanning electron microscope (Hitachi SU8020). Microscopic optical images were recorded using an optical microscope (Zeiss Observer Z1). XRD analysis was performed using an X'Pert3 Powder. Edinburgh FLS980 was used to measure the fluorescence spectrum and lifetime, while Shimadzu UV 3600 was applied for the analysis of the absorption spectra. The characterization of photoelectric performance was measured using an electrical testing system consisting of an adjustable power supply (Stanford DS 345) and a signal amplifier (Stanford SR 570). In pulse testing, green and red LEDs are both driven by

a DC power (Maynuo M8812). Informed consent was obtained from all human participants involved in the pulse testing and physiological signal monitoring experiments.

Results and discussion

Design strategy and device structure of the conformable PPG sensor

Fig. 1a illustrates the mechanism for the collection of PPG signals from humans. In daily life, the heart drives blood circulation through periodic contractions and relaxations, generating pulse waves during this process. Pulse waves propagate along the blood vessels to the whole body with the flow of blood. The signal of the pulse waves could be measured through the reflection and absorption of light in the skin, particularly at the fingertips which possess a lot of capillaries. The PPG signal consists of the direct current (DC) component and the alternating current (AC) component (Fig. 1b). The DC signal is determined by the static surface and tissues of the skin, muscle, and bone, for example, the dark skin results in a large DC baseline. The AC signal originates from the dynamic blood flow, which can be analyzed to obtain the physiological parameters, including heart rate, heart rate variability, and blood pressure.²⁵ Close contact between the PPG sensors and the skin is required for the accurate acquisition of the AC signals. The conformability of the PPG sensor is determined by its bending stiffness, which is usually proportional to the cube of the total thickness.^{26,27} We performed finite element analysis (FEA) simulations on the PPG sensors with the thickness ranging from $5 \text{ } \mu\text{m}$ to $125 \text{ } \mu\text{m}$, attached to a wavy surface that represents the skin texture (Fig. S1). The stress required to conform to the wavy surface increases drastically with the device thickness. The $5 \text{ } \mu\text{m}$ -thick device shows uniform stress distribution, ensuring good contact. Consequently, the ultrathin device can achieve a conformal contact with the skin (Fig. 1c), while the conventional flexible sensors often detach from the complex texture, leading to air gaps that introduce optical crosstalk (Fig. 1d).

Based on this strategy, the conformable PPG sensor was proposed with an ultrathin thickness of $5.2 \text{ } \mu\text{m}$. The device employs a sandwich structure where the 200 nm -thick perovskite film and the 30 nm -thick Au interconnects are encapsulated between two $2.5 \text{ } \mu\text{m}$ -thick parylene-C films, resulting in a total thickness of $5.2 \text{ } \mu\text{m}$ (Fig. 1e and Fig. S2). To realize this ultrathin freestanding device, it was first fabricated on a Si wafer for precise patterning and then released from the wafer using the electrochemical delamination process (see Experimental and Fig. S3).^{28,29} Fig. S4 shows the optical photographs of the device before and after stripping, confirming the unchanged morphology of the device. The entire sensor with an area of $2 \times 1.5 \text{ cm}^2$ weighs only 0.26 mg , corresponding to a density of 8.7 g m^{-2} (Fig. S5). The ultrathin properties enable the device to have excellent conformability to the textured surface. A flexible device with a thickness of $25 \text{ } \mu\text{m}$ was also fabricated with parylene-C films for comparison. We transferred these flexible and ultrathin devices to polydimethylsiloxane (PDMS) templates with sinusoidal roughness, where the amplitudes ranged from 0.5 mm to 1.0 mm and the

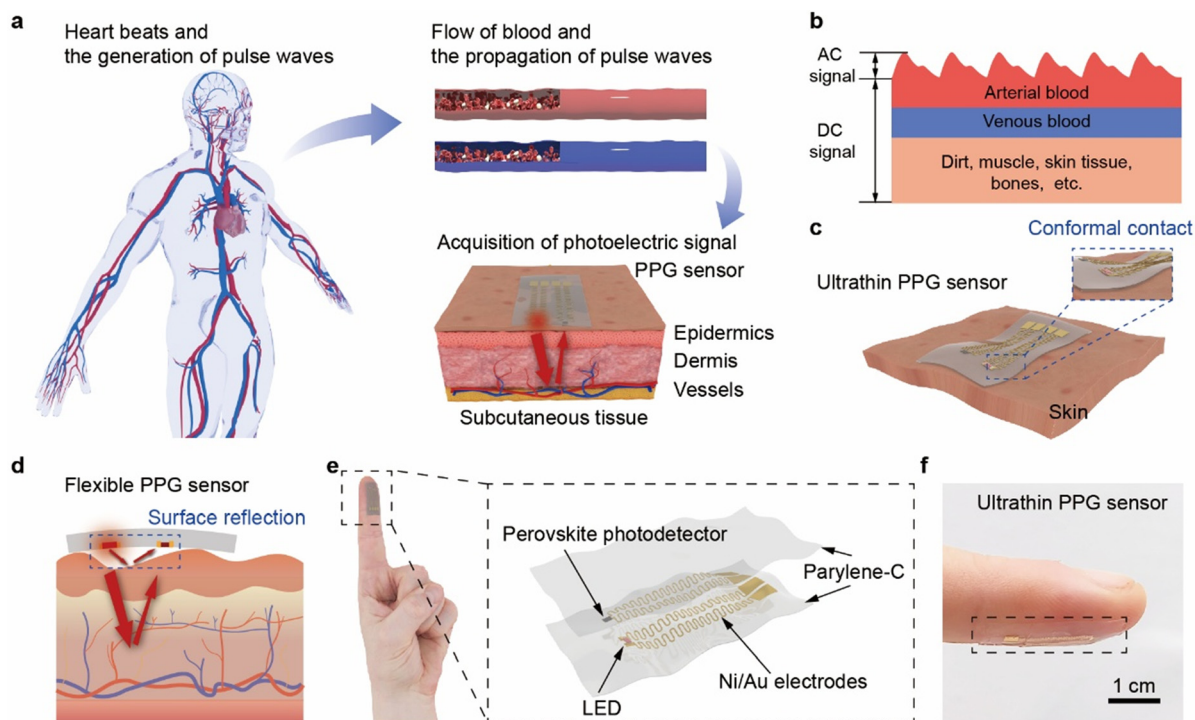


Fig. 1 Working principle and device structure of the conformal PPG sensor. (a) Schematic illustration showing the generation of pulse waves from the heartbeat, the propagation of blood flow, and the acquisition of photoelectric signals by the PPG sensor. (b) Schematic illustration of the compositions of the PPG signal. (c) and (d) Schematic illustration of the conformal PPG sensor with ultrathin structure forming conformal contact with the skin texture (c) and the flexible PPG sensor creating air gaps at the interface (d). (e) Device structure of the conformal PPG sensor. (f) Optical photograph of the ultrathin PPG sensor conformably attached to a human finger.

period ranged from 2.5 mm to 7.5 mm (Fig. S6). As shown in Fig. S7, the ultrathin device can adhere to the curved surface, replicating the template textures without detachment. However, the flexible device cannot bend to maintain contact, and it bridges across the peak positions due to the high rigidity. Furthermore, the ultrathin PPG sensor was applied to a fingertip (Fig. 1f). The device formed conformal contact with the finger, ensuring reliable optical coupling for accurate pulse signal collection.

Fabrication and characterization of the perovskite photodetector

To capture the weak reflected light from capillaries, a perovskite photodetector was employed in the ultrathin PPG sensor. A 2D RP perovskite $(\text{PEA})_2(\text{MA})_{n-1}\text{Pb}_n\text{I}_{3n+1}$ film was prepared by the surface energy-assisted crystallization method,^{30,31} ensuring precise alignment with the pre-patterned Au electrodes (Fig. S8, details in Experimental). Fig. 2a shows the scanning electron microscopy (SEM) image of the 2D perovskite film array, demonstrating that the 2D perovskite film can be fabricated in a designated location. The morphology of the perovskite film confirms its compact surface without pinholes (Fig. 2b). Fig. 2c shows the X-ray diffraction (XRD) pattern of the 2D perovskite film. Two obvious peaks at 14.2° and 28.5° were indexed to the (111) and (202) planes of the 2D perovskite, respectively. Furthermore, the energy-dispersive X-ray spectroscopy (EDS) mapping demonstrates uniform distribution of the Pb and I elements throughout the film (Fig. 2d). Optical properties of 2D perovskite film were characterized from

photoluminescence (PL), absorption, and time-resolution photoluminescence (TRPL) spectroscopy (Fig. S9). The PL peak is around 754 nm originating from the high- n phase composition of 2D perovskite. The absorption spectrum exhibits three obvious absorption peaks of ≈ 500 nm, 565 nm, 604 nm corresponding to 2D perovskite of $n = 1$, $n = 2$, $n = 3$ phase, while the absorption edge of 760 nm is attributed to the $n = \infty$ phase. PL peak and absorption edges were consistent with those of the previously reported 2D perovskite.^{32–34} The TRPL spectrum was fitted with a bi-exponential function, showing a lifetime of 2.58 ns for surface recombination and 13.88 ns for bulk recombination, respectively.

Based on the patterning method of the 2D perovskite film, the photoconductor-type photodetector was fabricated on an ultrathin substrate. As shown in Fig. 2e, the perovskite film was well aligned with the interdigital electrodes. Fig. 2f shows the representative current–voltage (I – V) curves of the photodetector under different illuminations. The dark current of the photodetector is 8.08×10^{-10} A, while the light current is 2.07×10^{-6} A under the illumination of 55.29 mW cm^{-2} , resulting in an on/off ratio of 2560. Moreover, the photodetector demonstrates stable switching behavior under both high and low illumination intensities (Fig. 2g). The dependence of photocurrent (I_{ph}) on the light intensity is plotted in Fig. 2h, which can be fitted by the equation:

$$I_{\text{ph}} = \alpha P^\beta$$

where P is the light intensity, and I_{ph} is the current difference between light (I_{light}) and dark (I_{dark}) conditions, expressed by

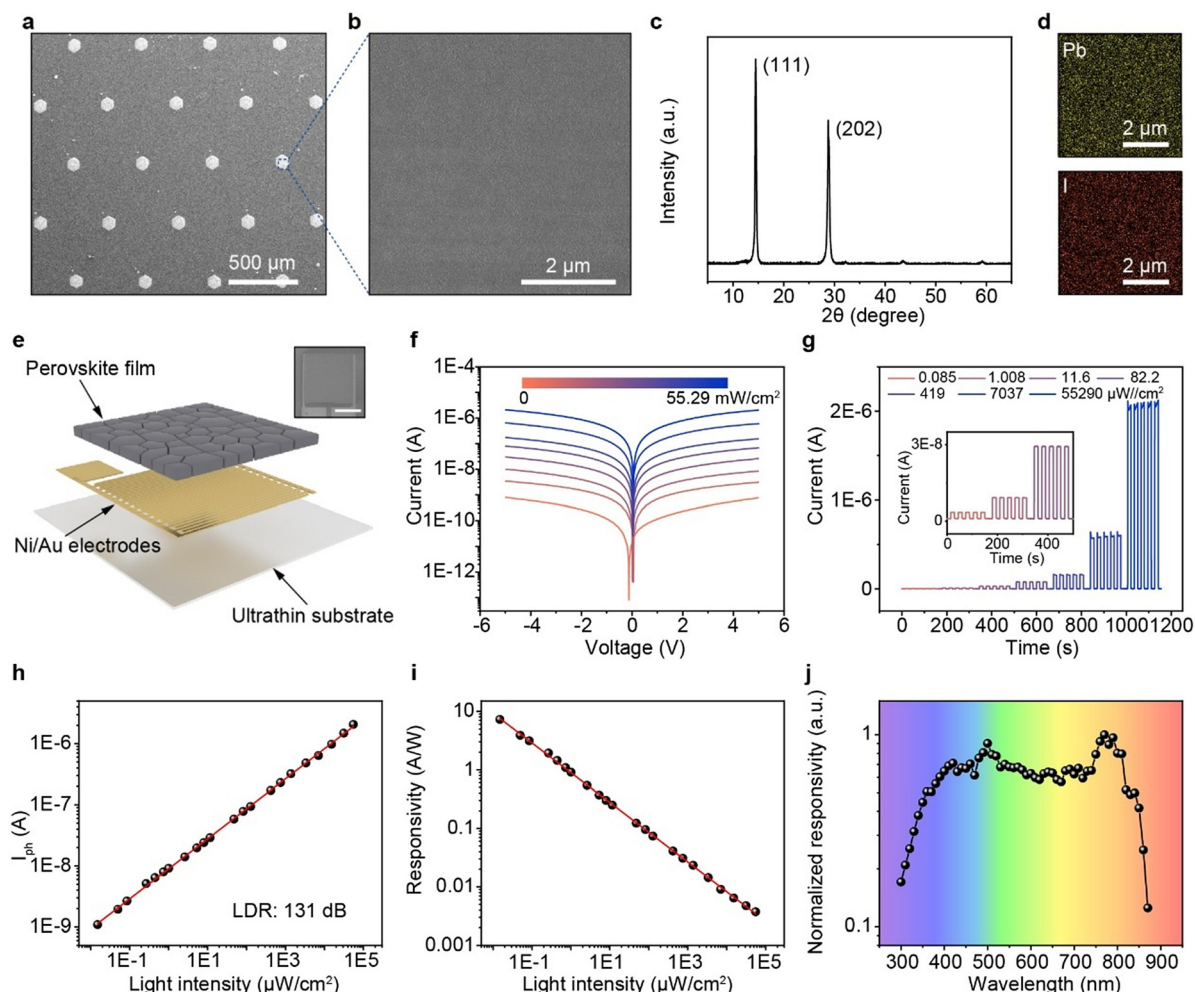


Fig. 2 (a) SEM image of the patterned 2D perovskite film array. (b) High-magnification SEM image showing the compact surface of the perovskite film. (c) XRD pattern of the perovskite film. (d) EDS mappings of Pb and I element distribution. (e) Schematic illustration of the perovskite photodetector structure. (f) and (g) I - V curves (f) and I - t curves (g) of the perovskite photodetector under 532 nm illumination with different light intensities. (h) and (i) The dependence of photocurrent (h) and responsivity (i) on the light intensity. (j) Normalized spectral responsivity of the perovskite photodetector across the visible to near-infrared region.

$I_{\text{ph}} = I_{\text{light}} - I_{\text{dark}}$.^{35,36} β is calculated to be 0.49, showing the sublinear relationship between I_{ph} and light intensity. In addition, the perovskite photodetector exhibits a large linear dynamic range (LDR) of 131 dB. Responsivity (R) is another important parameter of the photodetector, which can be calculated using the equation:

$$R = \frac{I_{\text{ph}}}{PA}$$

where P and A are the light intensity and working effective area, respectively.³⁷ As shown in Fig. 2i, the responsivity decreased gradually with the increase of light intensity. Under the illumination of 15 nW cm^{-2} , the photodetector reached a maximum responsivity of 7.25 A W^{-1} at a bias of 5 V. Fig. 2j exhibits the spectral response of the photodetector, covering a broad wavelength ranging from 400 nm to 800 nm. Furthermore, the response and decay time were calculated using the time intervals between 10% and 90% photocurrent peak values. As shown in Fig. S10, the response and decay time is 0.95 ms and 1.6 ms, respectively. This

fast photoresponse behavior enables the photodetector to resolve the dynamic features of the pulse waves.

Mechanical stability of the conformable PPG sensor

The mechanical stability of the device is crucial for reliable wearable PPG sensors. We first characterized the mechanical robustness of the optoelectronic performance of the perovskite photodetector. After the fabrication process of the device on the rigid Si wafer, the device was released in the NaCl solution (Fig. S3). As shown in Fig. 3a, both the dark current and photocurrent remained unchanged before and after the electrochemical stripping. The switching behavior was also maintained stable after stripping (Fig. 3b), confirming that the stripping process did not damage the sensitive perovskite layer. To characterize the mechanical tolerance of the PPG sensor, the device was fixed onto a bending stage, which can apply different bending angles (Fig. S11). Fig. 3c and Fig. S12 demonstrate the I - V curves of the perovskite photodetector with bending angles ranging from 0° to

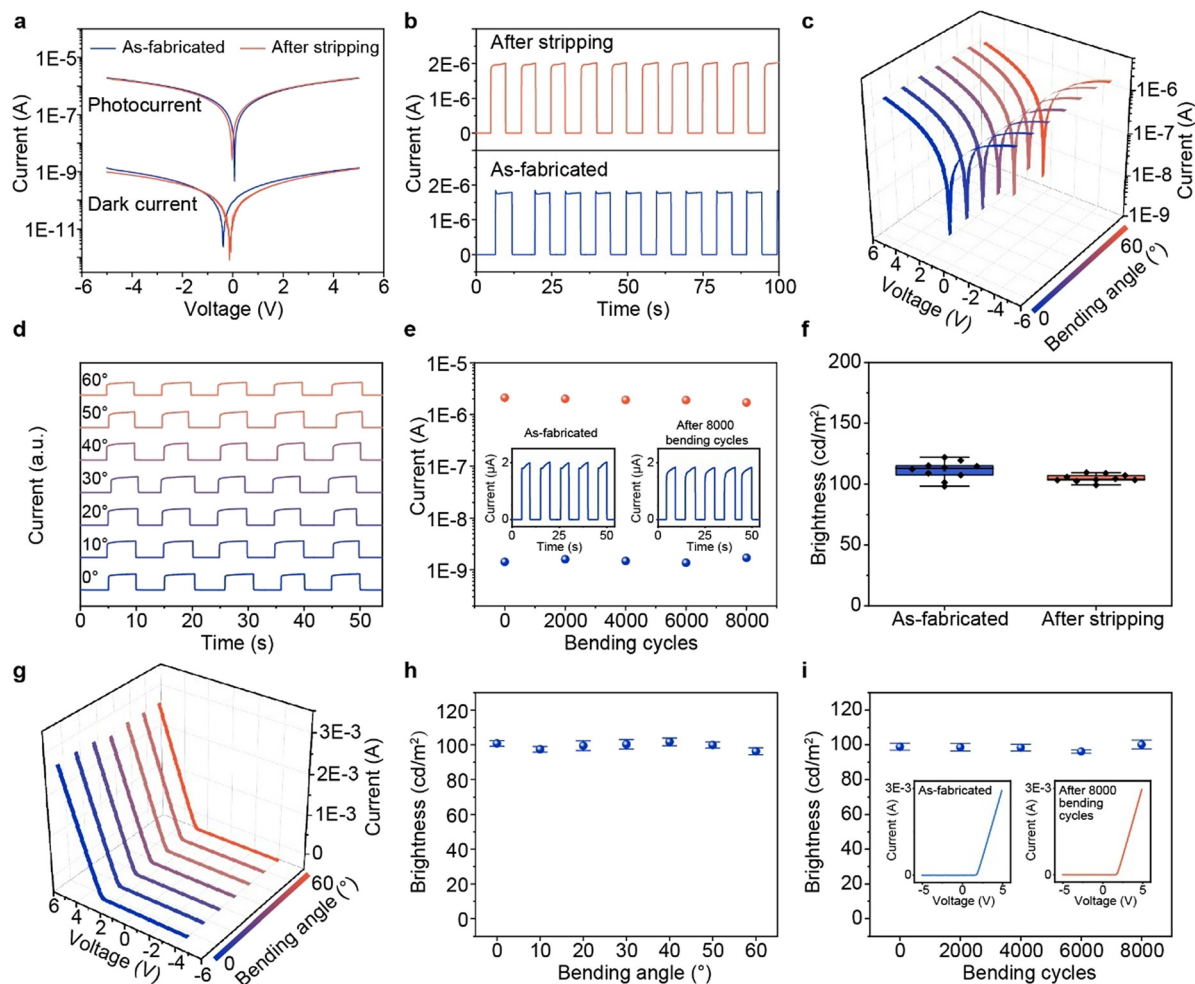


Fig. 3 (a) and (b) Comparison of the I - V curves (a) and I - t curves (b) of the perovskite photodetector before and after the stripping process of the PPG sensor. (c) and (d) Comparison of the I - V curves (c) and I - t curves (d) of the photodetector under different bending angles ranging from 0° to 60° . (e) Mechanical stability test of the photodetector showing stable current over 8000 bending cycles. The insets show the corresponding I - t curves of the photodetector as-fabricated and after 8000 bending cycles. (f) Statistical distribution of the LED brightness before and after the stripping process of the PPG sensor. (g) I - V curves of the LED under different bending angles ranging from 0° to 60° . (h) and (i) Brightness stability of the LED under different bending angles (h) and bending cycles (i). The insets in (i) display the I - t curves of the LED as-fabricated and after 8000 bending cycles.

60° under light illumination and dark conditions, respectively. These curves remained unchanged with varying bending angles. Furthermore, the switching behavior of the device also maintained a consistent on/off ratio and response time (Fig. 3d). A long-term bending test was also performed to investigate the bending durability of the device. As shown in Fig. 3e, both the dark current and photocurrent remained stable after 8000 bending cycles with a bending angle of 60° . These results confirm the excellent mechanical flexibility of the ultrathin device. In addition to the photodetector, the stability of illumination is also critical for the PPG sensor. Fig. 3f demonstrates the brightness of the LED before and after stripping. Although a slight decrease in brightness from 111.2 cd m^{-2} to 104.7 cd m^{-2} was observed, it is still suitable for PPG monitoring. The reliability of the electrical connection of the LED under deformation was investigated. As shown in Fig. 3g and h, the I - V curves of the LED remained constant under different bending angles, resulting in stable luminance with deformation. In addition, the

luminance was also stable after 8000 bending cycles with a bending angle of 60° (Fig. 3i). The I - V curves during the bending test further confirmed the robust electrical connection between the LED and Au electrodes (Fig. S13). These results demonstrate the reliable mechanical stability of the PPG sensor for wearable monitoring applications.

Accurate pulse acquisition and blood pressure sensing

To demonstrate the capability of the device for health monitoring, the ultrathin PPG sensor was worn on the index finger (Fig. 4a). The measurement circuit consisted of independent power supplies for the LED and the perovskite photodetector, and a signal amplifier was used to collect the current of the perovskite photodetector (Fig. S14). For comparison, the pulse signals were also collected using a flexible control device ($25 \mu\text{m}$ -thick). As shown in Fig. 4b, a large magnitude of the raw AC signal was observed in the ultrathin PPG sensor. The AC signals were extracted for further comparison (Fig. 4c). The ultrathin PPG sensor generated

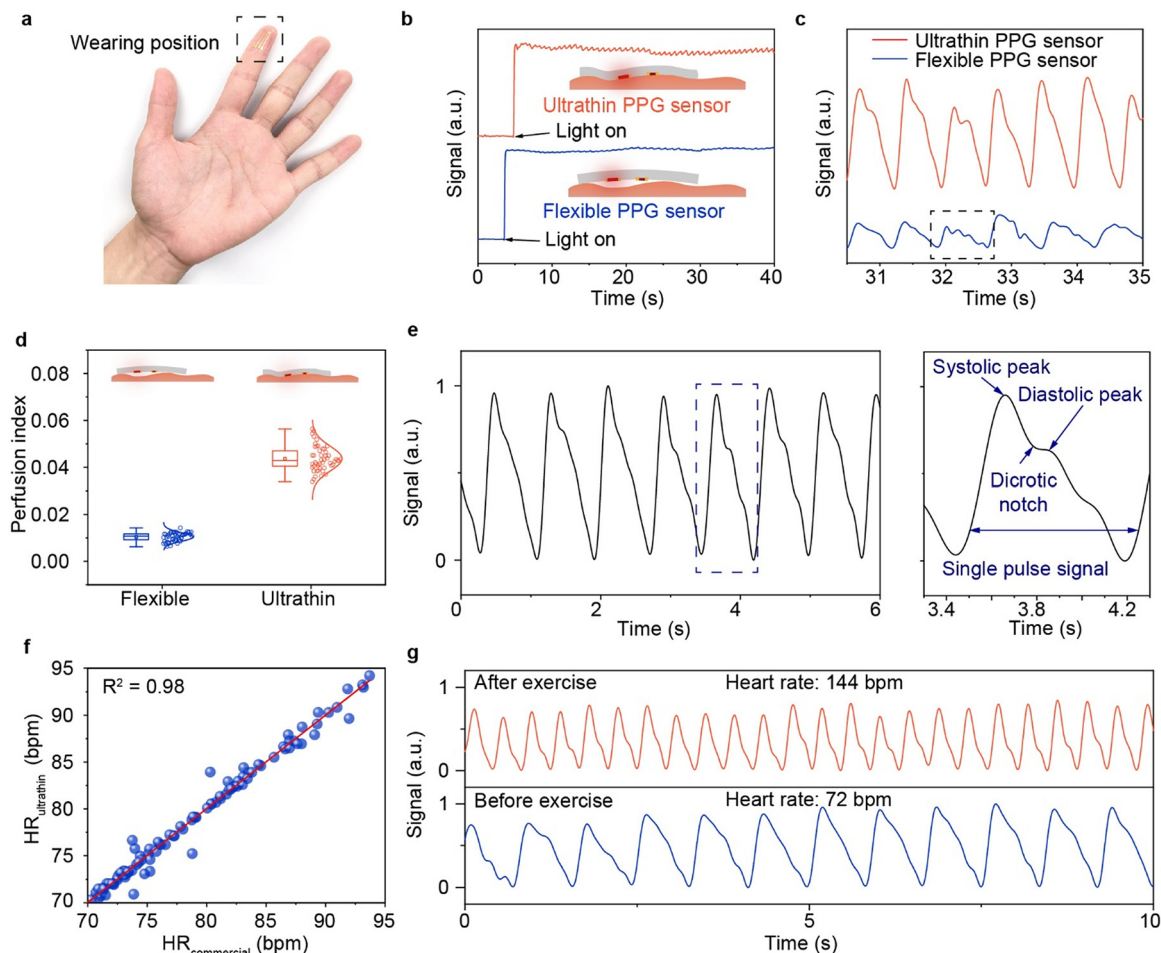


Fig. 4 (a) Optical photographs of the ultrathin PPG sensor attached to the index finger. (b) Comparison of the raw photocurrent response between the ultrathin and flexible control sensors during measurement. (c) Extracted AC pulse signals from both PPG sensors. (d) Statistical comparison of the perfusion index obtained from ultrathin and flexible PPG sensors. (e) A 6-second PPG waveform obtained from the ultrathin sensor. The right panel shows a magnified view of a single pulse cycle. (f) Correlation plot of HR measured simultaneously using an ultrathin PPG sensor and a commercial medical sensor. (g) Real-time monitoring of PPG signals and HR before and after exercise.

a strong and smooth pulse waveform with obvious systolic and diastolic peaks, whereas the signal from the flexible device demonstrated several noise peaks (black dashed box). The signal quality was evaluated using the perfusion index, which is defined as the AC/DC ratio. As shown in Fig. 4d, the ultrathin PPG sensor achieved a high perfusion index of 4.4%, which is about four times larger than that of the flexible PPG sensor. This enhanced perfusion index confirms the excellent signal coupling enabled by the conformal contact with the ultrathin structure. Fig. 4e demonstrates a pulse waveform for 6 seconds, where the characteristic peaks of the pulse, including the systolic peak, dicrotic notch, and diastolic peak, can be clearly resolved. The device also demonstrates reliable working stability, showing stable pulse recording for 180 seconds without noise and signal drift (Fig. S15). In addition, benefitting from the broadband spectral response and high sensitivity of the perovskite photodetector, the pulse waveform could be collected under both low-intensity red and green illuminations (Fig. S16). With the light intensity increasing, both AC and DC components increased linearly. However, the perfusion index is almost independent of the light intensity due to the

stable absorption properties of the human tissue, confirming the capability of pulse monitoring across a wide range of illumination intensities. It is worth noting that the device could also realize accurate pulse monitoring even under a low red light intensity of $25.3 \mu\text{W cm}^{-2}$ and a green light intensity of $36.7 \mu\text{W cm}^{-2}$. The accuracy of the device was validated against a commercial medical PPG sensor. The HR from our device was calculated according to the 5 pulse periods. Fig. 4f demonstrates the correlation of the HR measured simultaneously using both devices. A strong linear relationship with a correlation coefficient of $R^2 = 0.98$ was obtained. The Bland–Altman analysis was constructed to demonstrate the consistency of the two devices (Fig. S17). Most of the data points fall in the limits of the agreement ($\pm\text{LoA}$) with values of 1.73 and -1.65 , confirming the accuracy of our PPG sensor. We further monitored HR under different physiological states. As shown in Fig. 4g, the measured HR increased from 72 bpm (resting) to 144 bpm (after exercise).

Due to the accurate recording of the pulse waveform, the heart rate variability (HRV) could be analyzed. Fig. 5a demonstrates the Poincaré plot of HRV derived from the time

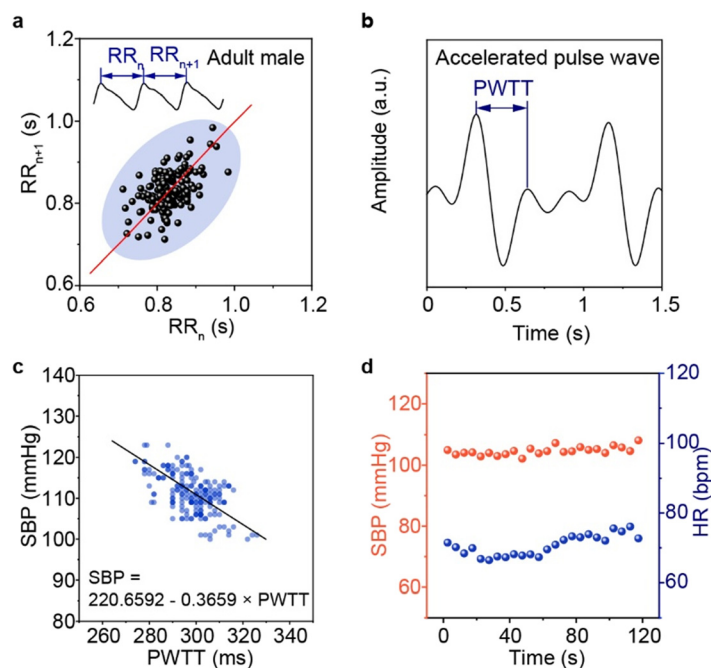


Fig. 5 (a) Poincaré plot of an adult for HRV analysis. (b) The accelerated pulse wave based on the original pulse wave measured using an ultrathin PPG sensor. (c) Correlation analysis between PWTT and SBP. The black line is the fitting line. (d) Continuous monitoring of SBP and HR for 120 s.

difference between adjacent peaks (R–R intervals), which exhibits a typical elliptical distribution for a healthy adult male. In addition, we explored the applications of cuffless blood pressure monitoring based on the accurate pulse waveforms.^{7,38}

The systolic blood pressure (SBP) was estimated using the pulse wave transit time (PWTT) method derived from the second derivative of the PPG signal (Fig. 5b and Supplementary Note 2 (see SI)). A linear relationship was then established between SBP and PWTT (Fig. 5c). The accuracy of this estimation method was validated with the Bland–Altman analysis, revealing a low mean bias of -3.6 mmHg and a standard deviation of 4.7 mmHg (Fig. S18). Based on this method, real-time monitoring of SBP and HR was achieved (Fig. 5d). Both SBP and HR signals remained stable, demonstrating the capability of the device for cardiovascular health monitoring.

Conclusions

In summary, we demonstrated a conformal PPG sensor for accurate pulse monitoring. Through the low-temperature fabrication process, the PPG sensor integrates a 2D perovskite photodetector and an LED using a parylene-C substrate and encapsulation, resulting in an ultrathin thickness of $5.2\ \mu\text{m}$ and a light weight of $8.7\ \text{g m}^{-2}$. This ultrathin architecture reduces device stiffness to enable conformal contact with the skin, eliminating the optical crosstalk at the interface. The photodetector demonstrates a high responsivity of $7.25\ \text{A W}^{-1}$ and a stable photoresponse under various mechanical deformations, enabling accurate pulse signal collection with a high perfusion index of 4.4% and a low operation intensity of $25.3\ \mu\text{W cm}^{-2}$. Furthermore, based on the high-fidelity pulse waveforms, HRV

analysis and continuous blood pressure monitoring were achieved. This work provides a straightforward method to enhance the pulse signal quality for accurate wearable health monitoring.

Author contributions

C. P. and X. H. supervised the research. C. P., X. H., X. P. and W. W. conceived the idea. X. H., W. W., and X. P. designed the experiments. X. P., X. H., W. W., S. J., M. G. and H. L. performed the experiments and analysed the data. Z. H., W. W., X. H., and C. P. wrote the paper. All authors discussed the results and commented on the manuscript.

Conflicts of interest

The authors declare no competing interest.

Data availability

The data supporting this article have been included as part of the supplementary information (SI). Supplementary information: experimental details on device fabrication, additional figures about material characterization studies, finite element analysis, and further data analysis related to the PPG sensor and blood pressure monitoring. See DOI: <https://doi.org/10.1039/d6tc00782a>.

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