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Self-Assembled Wavy Optical Microfiber for Stretchable Wearable Sensor

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Smart wearable devices have made remarkable success in medical-grade human vital signal monitoring and have promoted the realization of precision medicine and telemedicine. The flexibility and stretchability are significant for wearable devices, which determine the adhesion to the skin and influence the signal accuracy. Here, a stretchable and ultrathin optical sensor based on a self-assembled wavy microfiber is proposed. Thanks to the "bottom–up" strategy, the wave structure of the microfiber can be adjusted artificially, paving the way for the design of stretchable optical devices. Due to the microbend loss of the microfiber, the optical sensor has a sensitivity of ≈257 per unit strain and a good repetition (the standard deviation is 1.7% over 100 cycles). With the great sensor performance, the wrist pulse with a high signal-to-noise ratio is detected and the blood pressure is monitored successfully via the pulse arrival time, showing a potential application for human health monitoring.

1. Introduction

Over the past decades, flexible devices have made remarkable success in the application of smart wearable devices,[1,2] robots,[3,4] display,[5] energy,[6] and medical equipment.[7] Wearable sensors based on piezo-resistive devices, piezo-electrical devices, and capacitive devices have been employed to detect body motions,[8] pulse wave,[9] intracranial pressure,[10] and intraocular pressure,[1] realizing the human health monitoring and biomedical applications.

The recent development of electronic wearable sensors triggered the research of photonic sensors based on optical fiber. Li et al. proposed a sensitive optical microfiber sensor based on a hybrid plasmonic microfiber knot resonator, realizing the detection of the pulse wave.[11] Zhang et al. proposed the highly sensitive skin-like wearable optical sensors and realized the detection of the wrist pulse and the human voice.[3] Pan et al. proposed a multifunctional skin-like wearable optical sensor that realized the detection of respiration and body temperature.[12] With the advantages of the electromagnetic immunity, high sensitivity, and fast response time, wearable optical sensors can be one of the best options to obtain high signal-to-noise ratio physiological signals. Besides, sensors based on optical fiber can easily construct networks. Distributed sensing can be realized to provide more information about the body status, which is significant for future real-time health monitoring and the realization of telemedicine and precise medicine.

When it comes to wearable devices, flexibility and stretchability are of vital importance. Materials and structure designs are two core contents to realize devices stretchable. Optical microfibers with a waist diameter from several hundred nanometers to several micrometers owe extreme flexibility and configurability. However, the brittle material of the microfiber (silica) limits the stretchability of the optical device. So, it is challenging to make optical devices stretchable and stable. Pan et al. prebent the optical micro-/nano-fiber, realizing the sensor stretchable.[12] However, patterning methods that are suitable for large-scale production still need to be discovered for stretchable optical devices. Besides, optical sensors should become thinner to have a faster response and increase the comfort level when people wear them.

In this work, we report a stretchable and ultrathin optical sensor based on a wavy microfiber. The approach inspired by the "bottom–up" strategy for forming wavy geometries self-assembly in microfibers is controllable and suitable for mass production, which is different from the patterning methods of the microfiber in previous articles[3,12] and paves the way for the design of the stretchable optical devices based on the microfiber. Embedding the wavy microfiber in an ultrathin polydimethylsiloxane (PDMS) film, a flexible sensor is realized for human health monitoring. The wrist radial artery was chosen as the measurement site and the optical sensor was stuck on it by Van der Waals force. With the sensor's sensitivity and stability, the pulse wave with a high signal-to-noise ratio was detected. And then with the comparison with the electro-cardio signal (ECG), human blood pressure was successfully monitored via the pulse arrival time (PAT), showing a potential for the real-time health care application.

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2. Design and Fabrication

Figure 1 shows the schematic illustration and photographs of the optical sensor based on the wavy microfiber. The microfiber was tapered by a standard commercial single-mode fiber (the cladding diameter of 125 µm and the core diameter of 9 µm) by the flame brushing method.[13] The microfiber with a diameter of several hundred nanometers to several micrometers and the wave structure was encapsulated into a thin PDMS film, which has a thickness of ten micrometers to several hundred micrometers, realizing the flexible and stretchable optical sensor. Figure 1b shows the optical sensor wrapped on a glass tube with a red light passing through it, revealing the sensor's flexibility. As shown in Figure 1c, the thickness of the optical sensor was similar to an A4 sheet, which guarantees the mechanical response and the adhesion to the human skin by Van der Waals force.

The approach for forming wavy geometries in microfibers was inspired by the "bottom–up" strategy, which has been employed to form the buckling structure of electrodes to realize the devices stretchable in flexible electronics.[14,15] The fabrication process of the optical sensor is shown in **Figure 2**a. First, the microfiber was tapered by a standard commercial single-mode fiber by the flame brushing method.[13] The diameter of the microfiber that we taped can be controlled from several hundred nanometers to tens of micrometers. Then, the microfiber was glued by the PDMS liquid prepolymer on a thin PDMS film (Base: Curing agent = 10:1), which was elastically stretched in advance. The strain in the PDMS film was released and the microfiber was deformed into the wave structure in the horizontal direction due to the cylindrical structure. At last, the wavy microfiber was embedded by brushing a thin layer of PDMS, followed by heating to 80 °C for 2 h.

Apart from the elastic modulus and the Poisson's ratio of microfiber and the PDMS film, the structure of the wavy microfibers is also determined by the waist diameter of the microfiber and the prestrain of the PDMS film.[14] For example, as shown in Figure 2b, when the prestrain of the PDMS film was fixed at 4%, the wavelength and amplitude of the microfiber's wave structure were both increased linearly with the waist diameter. The pattern was sinusoidal. The wavelength was between 349 and 977 µm and the amplitude was between 22 to 52 µm, depending on the waist diameter of the microfiber. And in Figure 2c, when the prestrain was increased from 4% to 10% and the waist diameter was fixed (5 µm for these data), the wavelength decreased slightly and the amplitude increased. The behavior in this static wavy configuration of the microfiber is similar to the wavy structure of the membrane, which is consistent with the nonlinear analysis of the initial buckled geometry in a uniform, thin, high-modulus layer on a semi-infinite low-modulus support.[14] The dynamic regulation of the wave structure is quite important for stretchable devices because it determines the stretchability and the effective modulus of the device.[16]

3. Results and Discussion

3.1. Characterization of Stretchable Optical Sensor

Thanks to the wave structure of the optical microfiber, the sensor was endowed with stretchability. The maximum tension range that we realized was 10%, which was dependent on the prestrain of the PDMS film in the fabrication process. The stretchability was largely increased compared with the normal optical fiber whose maximum strain is ≈1%. What's more, the optical sensor is sensitive to weak strains due to the microbend loss caused by the wavy structure.

In order to investigate the strain response, a typical stretchable optical sensor with a length of ≈21 mm was attached to a translation stage. The waist diameter of the microfiber was 9 µm. The wave structure was 3 mm in length, and its amplitude and wavelength were about 22 and 480 µm separately. A 1550 nm fiber laser with stable light power was employed as the light source and the output power was recorded by a dynamometer (Thorlabs, S145C). **Figure 3**b,c shows the structure change

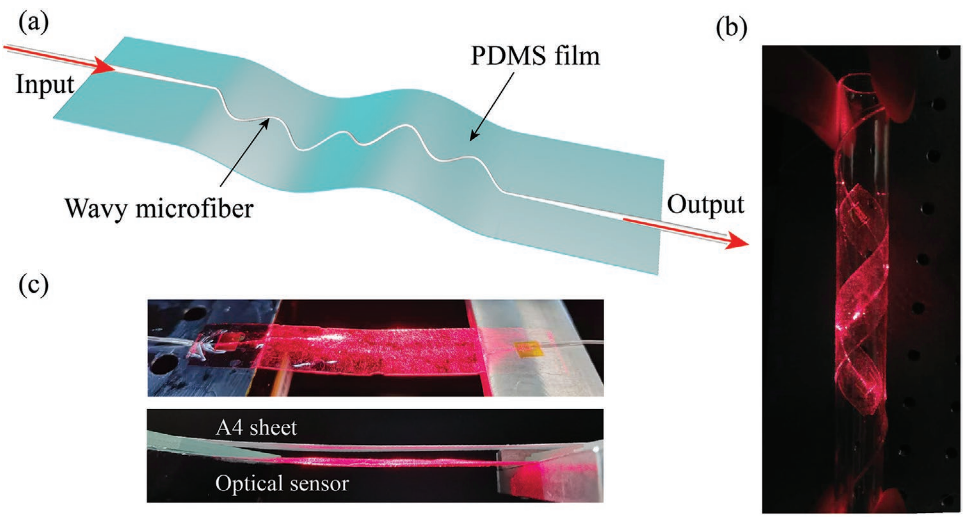

Figure 1. a) Schematic diagram of the stretchable optical sensor. b) Photograph of the flexible optical sensor wrapping on a glass tube. c) Photograph of the optical sensor suspended in air and compared with an A4 sheet.

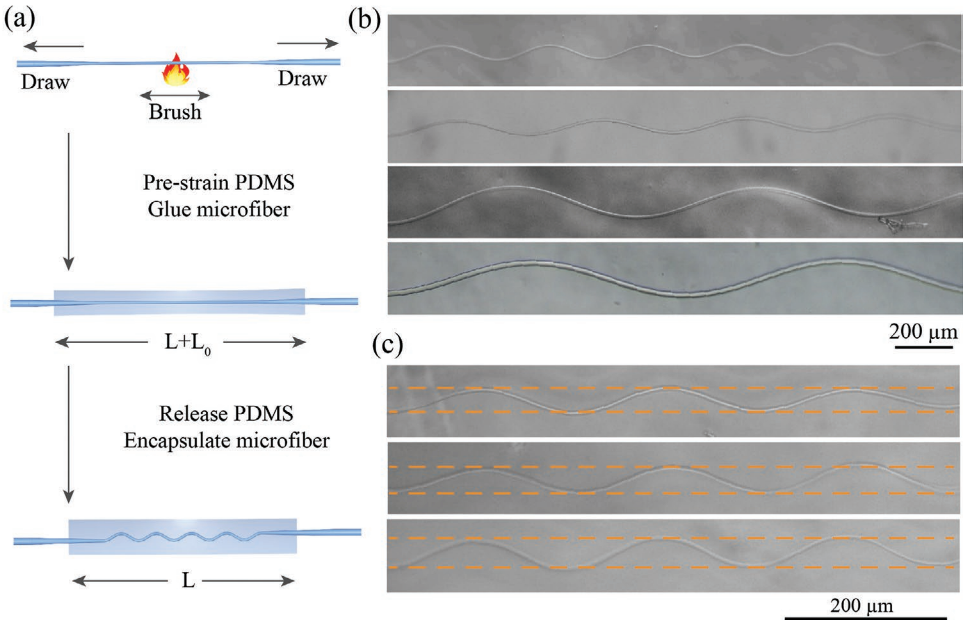

Figure 2. a) Fabrication process of the optical sensor. b,c) Photographs of the wave structure with b) different waist diameters of the microfiber and c) different prestrain of the PDMS film.

and the response of the optical sensor when stretched. The wave structure flattened out and the transmittance increased due to the decrease of the microbend loss.[12] So, the strain sensitivity of the optical sensor is dependent on the length and curvature of the wave structure and the waist diameter of the optical microfiber to a large extent. The optical gauge factor, G,[11] is employed to characterize the strain-sensing performance of the stretchable optical sensor:

$$G = (\Delta T/T)/\varepsilon = (\Delta P/P)/\varepsilon \quad (1)$$

where $\Delta T/T$ and $\Delta P/P$ are the relative transmittance change and relative output power change, respectively. ε is the applied strain on the sensor. The results show that the optical sensor owns a gauge factor of ≈257. And the sensitivity in unit length (=G/L, where L is the length of the wave structure) is 85.7 mm^{-1}, which is better than the stretchable strain sensor based on the prebent optical microfiber.[12] With great sensitivity, the minimum strain that can be detected is 9×10^{-5} considering the noise level of the system. The reversibility of the optical sensor was further investigated by automatically stretching and releasing the sensor about 100 cycles using a computer-controlled translation stage. The standard deviation was ≈1.7% over 100 cycles and the baseline intensity was quite stable.

3.2. Pulse Wave Detection and Data Processing

With great sensitivity and reversibility, the stretchable optical sensor can be employed to detect the human's pulse wave via recording the expansion of the vessel. A 20 μm thin flexible sensor with a 2 μm microfiber was stuck on the wrist by Van der Waals force with good contact, which is shown in **Figure 4**a. A 1550 nm fiber laser with stable light power was employed as the light source. The output power was transformed into the electrical signal via a photodetector and was recorded by an oscillometer.

As shown in Figure 4b, the amplitude of the raw pulse wave signal was ≈5 mV, and the amplitude of the high-frequency noise was ≈0.1 μV. So, a pulse waveform with a high signal-to-noise ratio of ≈94 dB was detected. Apart from the pulse wave, because of the sensing mechanism based on the light intensity, the waveform involved the high-frequency noise and the baseline drift caused by motion interference. So, a signal processing algorithm was employed to extract the pulse wave. First, a low-pass filter with a cutoff frequency of 15 Hz was used to filter high-frequency noise. Then the baseline was extracted via open–close operation based on the mathematical morphology.[17] Figure 4b shows the raw waveform, the baseline waveform, and the filtered waveform. Results show that after signal processing the pulse signals presented better uniformity, which lay a foundation for follow-up data analysis. Besides, the baseline waveform correlated with the respiration due to the slight fluctuation of the arm caused by the breath.[11]

As shown in Figure 4c, thanks to the sensitivity of the optical sensor, the details of the pulse wave, such as main peak, tidal-wave, dicrotic notch, and dicrotic wave, can be distinguished vividly. The pulse cycle can be divided into the systolic phase and the diastolic phase by the dicrotic peak. The waveform of the pulse wave (for example, the time delay and the relative amplitude of the tidal-wave and dicrotic peak) tells the health state of the cardiovascular.

3.3. Pulse Arrival Time Detection and Blood Pressure Monitoring

After data processing, the pulse signal with a high signal-to-noise ratio was employed to monitor the blood pressure via the comparison with the ECG.[9,18] **Figure 5**a shows the positions of the optical sensor and ECG electrodes. As shown in Figure 5b,c,

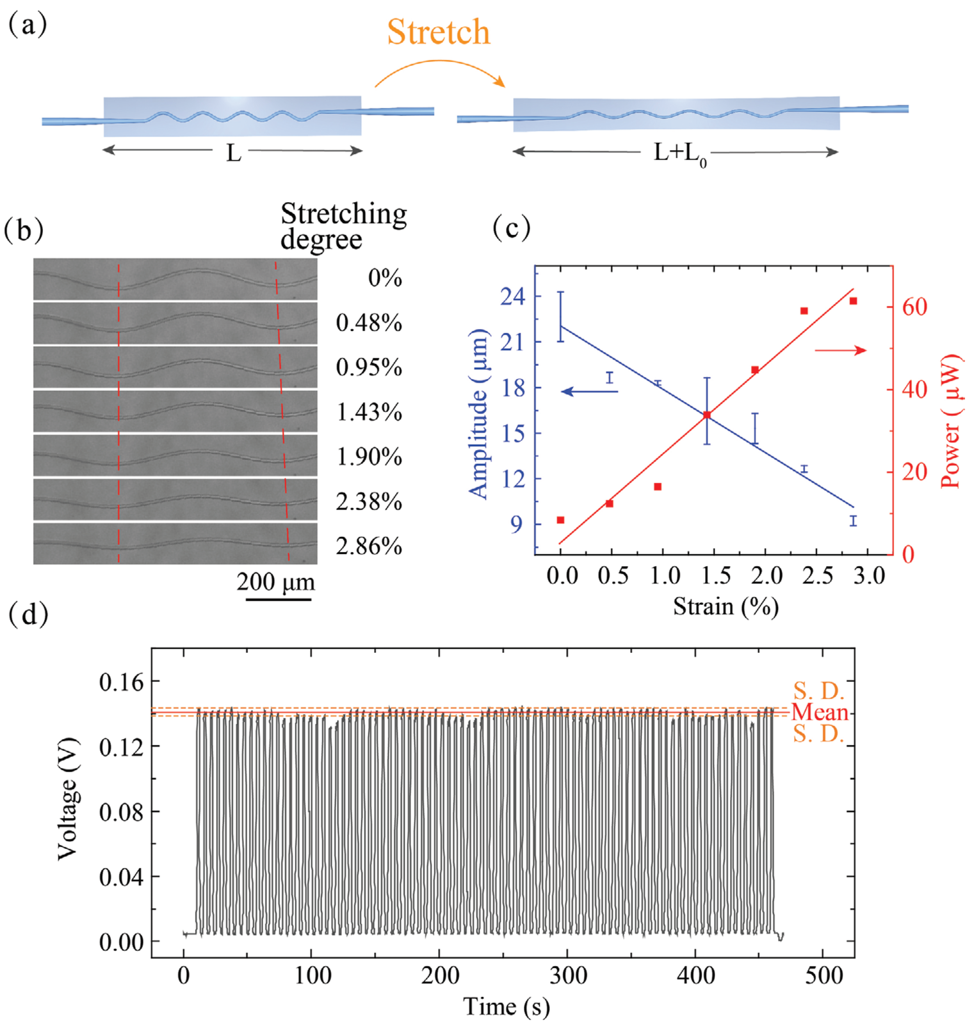

Figure 3. a) Schematic diagram of the strain response characterization. b) Structure and c) output power of the sensor under different stretches. d) Reversibility test for about 100 cycles.

the time delay, which was defined as the PAT, was calculated from the R-peak of the ECG waveform collected by the AD8232 heart rate monitor and the systolic peak of the pulse waveform collected from the wrist by our optical sensor.

According to the Moens–Kortweg equation, the PAT is related to the elasticity of the artery:[19]

$$\mathrm{PWV} = K\sqrt{\frac{hE}{\rho d}} \tag{2}$$

$$\text{Time Delay} = \frac{L}{\mathrm{PWV}} \tag{3}$$

where PWV represents the pressure wave velocity and the artery is assumed to be an elastic tube with a thickness h, diameter d, and blood density ρ. Moens constant K is 0.8 for the human's central artery. The elastic modulus (E) of the vessels is related to the pressure P as below:

$$E = E_0 e^{\alpha P} \tag{4}$$

where α is a vessel parameter and E_0 is the elastic modulus for zero pressure. Thus, by detecting the PAT, the variation of the individual blood pressure was monitored.

As shown in Figure 5d, the blood pressure fluctuation of a 25-years-old healthy man during 3 days was monitored by our cuffless blood pressure system and a cuff-based sphygmomanometer as a calibration. At each measurement, the pulse wave and the ECG signal were recorded for 30 s. The R-peak of the ECG waveform and the systolic peak of the pulse waveform were extracted to calculate the PAT. At the same time, the blood pressure was calibrated by the cuff-based sphygmomanometer thrice. During the 3 days, the systolic pressure had a fluctuation from 116 to 128 mmHg and the PAT changed from 303.3 to 296.7 ms. The PAT and the blood pressure own a similar trend, which proves the potential of our optical sensor for human health monitoring.

4. Conclusions

In conclusion, we have demonstrated a stretchable and ultrathin optical sensor based on a self-assembled wavy microfiber.

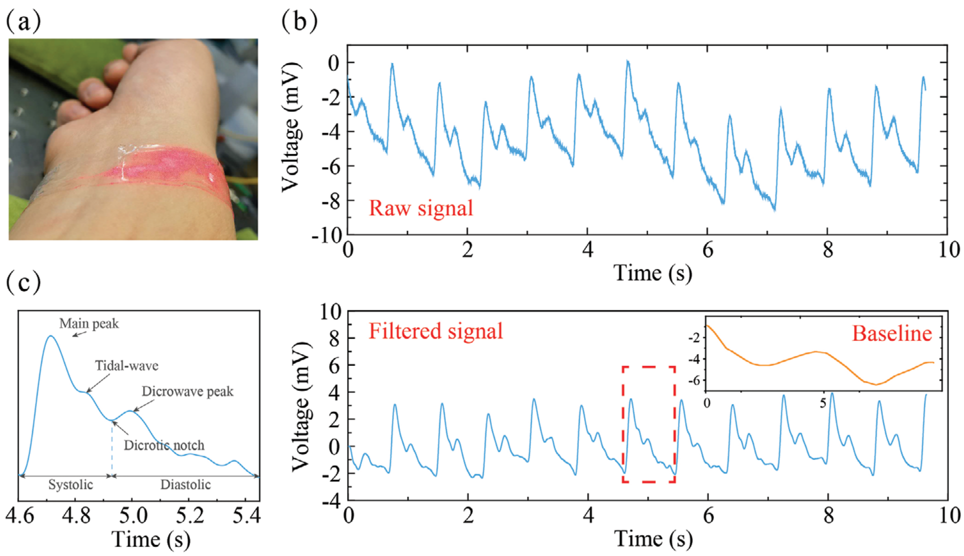

Figure 4. a) Photograph of the flexible optical sensor stuck on the wrist. b) Raw waveform, baseline waveform, and filtered waveform of the pulse wave. c) Details of the pulse wave.

Thanks to the "bottom–up" strategy, the wave structure of the microfiber can be adjusted artificially via the control of the waist diameter of the microfiber and the prestrain of the PDMS film. The fabrication method paves the way for design of the stretchable optical devices based on microfiber. Due to the microbend loss of the microfiber, the optical sensor has a sensitivity of ≈257 per unit strain. Besides, the optical sensor also has a good repetition and the standard deviation of the output voltage is ≈1.7% over 100 cycles. The wavy microfiber was embedded into a thin PDMS film, which provided an excellent adhesion to the skin via Van der Waals force and guaranteed the response time of the optical sensor. The pulse wave with a high signal-to-noise ratio was detected by sticking the optical sensor on the wrist. With the help of signal processing, a clean pulse waveform was obtained stably. According to the waveform, several parameters about human's vital signs and cardiovascular status can be monitored, such as the heart rate (HR), the stiffness index (SI), and the reflection index (RI).[11,20] What's more, the optical sensor can be employed under extreme environments owing to the electromagnetic immunity of the optical fiber. For example, patients' vital signs can be monitored under magnetic resonance imaging in the hospital. And then, the blood pressure was dynamically tracked via the PAT calculated by the pulse and ECG waveform, revealing the ability of daily human health monitoring.

5. Experimental Section

Fabrication of Wavy Microfiber Sensors: First, the microfiber was tapered by a standard commercial single-mode fiber (cladding diameter of 125 µm and core diameter of 9 µm) by the flame brushing

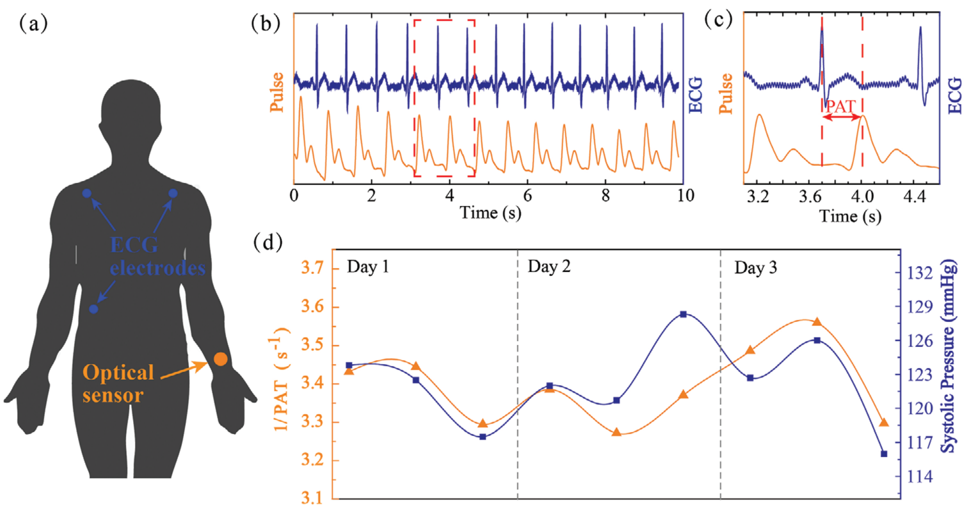

Figure 5. a) Schematic diagram of the blood pressure monitoring. b) Waveform of pulse and ECG collected at the same time. c) Schematic diagram of the PAT calculation. d) Blood pressure monitoring based on PAT detection.

method. A hydrogen flame was used to heat the fiber to its softening temperature. And then, the optical fiber was stretched by two high-precision translation stages at a velocity of 0.2 mm s^{-1}. The diameter of the microfiber that the authors taped could be controlled from several hundred nanometers to tens of micrometers depending on the shifting distance of the translation stages and the flame's brush range. Then, the microfiber was glued by the PDMS liquid prepolymer on a thin PDMS film (Base: Curing agent = 10:1), which was elastically stretched in advance. The strain in the PDMS film was released and the microfiber was deformed into the wave structure. At last, the wavy microfiber was embedded by brushing a thin layer of PDMS (≈5 µm in thicknesses), followed by heating to 80 °C for 2 h.

The optical sensors used to explore the wavy structure had a waist diameter of the microfiber from 5 to 20 µm and the prestrain of the PDMS film from 4% to 10%. The optical sensor with a 9 µm waist diameter and 4% prestrain was employed to characterize the structure variation, sensitivity, and reversibility under different strains. The optical sensor with a 2 µm waist diameter and 4% prestrain was employed to detect the pulse wave and monitor the blood pressure.

Characterization of Sensitivity and Reversibility: For characterizing the sensitivity of the optical sensor, a typical stretchable optical sensor with a waist diameter of ≈9 µm was attached to a translation stage. A 1550 nm fiber laser with stable light power was employed as the light source and the output power was recorded by a dynamometer (Thorlabs, S145C). The strain was applied to the optical sensor via moving the translation stage. Photographs of the wave structure were taken by the microscopy at the same time.

For characterizing the reversibility of the optical sensor, the optical sensor was attached to a computer-controlled translation stage. A 1550 nm fiber laser with stable light power was employed as the light source. The output power was transformed into the electrical signal via a photodetector and was recorded by an oscillometer. The translation stage stretched the optical sensor 100 times with the same shifting distance. The peak voltages of each stretch were extracted to calculate the mean value and standard deviation to evaluate the reversibility of the optical sensor.

Pulse Wave Detection and Blood Pressure Monitoring: To detect the pulse wave, a 25-years-old healthy man sat quietly. A 20 µm thin flexible sensor with a 2 µm microfiber was stuck on the wrist by Van der Waals force with good contact. A 1550 nm fiber laser with a stable light power of 30 mW was employed as the light source. The output power was transformed into the electrical signal via a photodetector and was recorded by an oscillometer. The sampling rate was 1 kHz.

To process the pulse waveform, a Butterworth low-pass filter with a cut-off frequency of 15 Hz was employed to filter high-frequency noise. And then, a linear structural element with a length of 400 sampling point was employed to extract the baseline signal via the mathematical morphology.

When monitoring the blood pressure, the wrist pulse wave and the ECG signal were recorded for 30 s synchronously by an oscillometer with a sampling rate of 1 kHz. The above signal processing algorithm was employed to filter the pulse wave. The R-peak of the ECG waveform and the systolic peak of the pulse waveform were extracted to calculate the PAT. At the same time, the blood pressure was calibrated by the cuff-based sphygmomanometer thrice.

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Conflict of Interest

The authors declare no conflict of interest.

Author Contributions

F.X. and Q.D. conceived the project idea and designed the experiments. H.-T.Z., L.-W.Z., Y.C., and Q.D. carried out the experiments and collected the data. H.-T.Z., L.-W.Z., and Q.D. analyzed all the data. H.-T.Z., Y.C., and F.X. cowrote the paper. All authors discussed the results and commented on the manuscript.

Data Availability Statement

The data that support the findings of this study are available from the corresponding author upon reasonable request.

Keywords

blood pressure monitoring, optical microfibers, pulse wave detection, self-assembled wave structures, stretchable optical sensors

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