Flexible and Printed Microwave Plasmonic Sensor for Noninvasive Measurement

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\textbf{ABSTRACT} The printed electrodes for detecting direct current signal changes of human vital signs have been fully investigated. Here a flexible and printed microwave plasmonic sensor for detecting liquid solutions is proposed and demonstrated. The sensor for noninvasive measurement at microwave frequencies is based on the spoof localized surface plasmons resonator, which is composed of a metal corrugated ring fabricated on the flexible PET substrate using the inkjet printing technology. The winding-shaped polydimethylsiloxane (PDMS) microfluidic channel is interleaved with the corrugated ring to simulate the complicated blood vessel. The simulated results agree well with the experimental measurements. It shows that the measured resonance frequency offset of 147 MHz has been achieved when the microfluidic channel is filled with deionized water, which indicates that the flexible microwave biological sensor is feasible. The calculated figure of merit is as high as 1178. The sensor can find wider applications in the flexible and wearable device field for continuous health monitoring.

\textbf{INDEX TERMS} Flexible printed circuits, microfluidics, plasmons, wearable sensors.

\section{I. INTRODUCTION}
Flexible electronics possess the advantages of light weight, high flexibility, great conformability, low manufacturing costs by use of inexpensive plastic substrate materials and high-volume manufacturing scalability through roll-to-roll processing [1], [2]. Flexible sensing electronics will change conventional diagnosis methods and revolutionize medical instruments with portable, wearable, remote, and timely features [3]. Over the last few decades, a wide range of revolutionary functional devices have been demonstrated, including flexible transistors [4], [5], flexible carbon nanotube integrated circuits [6], smart sensors [7], roll-up displays [8], electronic skin [9], [10], etc. Driven by the increasing demands for continuous health monitoring, daily and sports activity tracking, and multifunctional electronic skin, there has been a need for wearable devices and sensors [11]. Flexible medical devices designed for both fitness monitoring and medical diagnostics have been reviewed, where human vital signs, such as body temperature, heart rate, respiration rate, blood pressure, pulse oxygenation, and blood glucose are monitored [12]. An ideal platform for personalized wearable devices based on flexible electronics has been demonstrated, where human sweat, tears, saliva, blood, interstitial fluids, wound fluids as well as exhaled breath are continuously and non-invasively monitored [13]. Wearable and attachable health monitoring systems have been proposed to improve the comfortability and real-time health monitoring in clinical diagnoses and in daily life [14], [15]. By using a set of soft, flexible, and stretchable electronic devices, the concept of a “lab-on-skin” which conformally contact with the epidermis to deliver a range of functionalities has been introduced [16].

The manufacturability of flexible electronics is significantly enhanced by encompassing various printing technologies such as inkjet printing [17], electrospinning [18], transfer printing [19], screen printing [20] and gravure printing [21]. A printed, flexible, wearable lactate sweat sensor was demonstrated to non-invasively monitor sweat in real-time [22]. Direct printing can be done on a variety of disposable substrates, which is also compatible with flexible substrates with form factors closely matching that of skin. Furthermore, it is a high-throughput additive process leading...
to inexpensive products which are necessary in the case of disposable sensors [22]. Using the direct printing technique, a wearable sensing patch which is capable of measuring electrocardiography (ECG) signal and skin temperature was fabricated [23]. Printed and flexible sensors are suitable for monitoring vital signs as well as analytes in bodily fluids, due to the high-fidelity sensor-skin interfaces [12], [24], [25]. Most of previous works use the printed electrodes to detect the small direct current (DC) electrical changes of the body. In the last few years, the idea of developing sensors based on microwave detection and microtechnology for biological characterization has been emerging [26], [27]. The characteristic parameters of the fluids, even very small volumes (lower than few µL), can be accurately determined based on the measured impedance, resonant frequency, insertion loss, or the phase of the microwave or radio frequency (RF) signals [15], [28]–[35], with the advantages of robustness, light weight and durability. However, the sensitivity is still limited.

Periodically textured closed surfaces can support spoof localized surface plasmons (LSPs) at microwave and terahertz (THz) frequencies, which are sensitive to the changes of the surrounding environment and called plasmonic metamaterials [36]. Multipolar spoof LSPs on the ultrathin textured metallic disk were demonstrated experimentally at microwave frequencies, providing higher sensitivity than the dipole mode [37]. By introducing a ground plane underneath the textured metallic disk, higher-order spoof LSPs with enhanced quality factors have been demonstrated [38]. Spoof LSPs on an ultrathin metal-insulator-metal ring resonator were experimentally shown with increased quality factors for higher sensitivity [39]. In this work, we present a flexible and printed microwave plasmonic sensor to non-invasively and accurately detect high-loss polar solutions, such as the blood. The spoof plasmonic sensor is printed on the flexible substrate through the inkjet printing technology. The high-loss polar material under test is filled in the PDMS microfluidic channel to simulate the blood vessel. When the channel is empty, the measured resonance frequency is 5.303 GHz, and it changes to 5.156 GHz, with a frequency shift of 147 MHz, when the microfluidic channel is filled with deionized (DI) water. The sensor can find wider applications for the flexible and wearable devices.

II. FLEXIBLE SPOOF PLASMONIC RESONATOR

The polydimethylsiloxane (PDMS) elastomer is widely used in microfluidic sensing for many advantages: flexible, bio-compatible and durable, etc. Here we use it to simulate the skin issue which consists of the epidermis, the fat and the blood. Here the PDMS microfluidic channel is equivalent to the blood vessel and the injected liquid is equivalent to the blood, as shown in Figure 1. The flexible sensor is composed of the inkjet-printed plasmonic resonator on the flexible PET film. The PET membrane is bonded to the surface of PDMS using adhesive film (ARcare 92848), which is reliable and bio-compatible [34].

Figure 2(a) shows the detailed geometry of the spoof plasmonic sensor which consists of the top corrugated metallic ring and the flexible dielectric substrate (PET). The patterned metallic layer is a spoof plasmonic resonator, composed of closed corrugated strip with periodic array of grooves, where the number of the grooves, the inner radius, the groove height and the strip width are $N = 40$, $r = 6$ mm, $h = 6$ mm, and $g = 0.8$ mm, respectively. The 50 Ω microstrip transmission line and the impedance matching part are adopted to excite the spoof plasmonic resonator. The length $l$ of the microstrip line is set to be 6 mm. The outer radius of the arc for impedance matching is designed as $r_{arc} = r + g + h + w$, where $w = 0.5$ mm denotes the width of the circular arc. The three-dimensional (3D) perspective view of the sensor has been illustrated in Fig. 2(b), and the physical size of the resonator is 40 mm $\times$ 32 mm. The angle $\phi$ of the metal matching circular arc is 23°. The thickness $t_{s}$ of the PDMS elastomer is 2.5 mm. The relative dielectric constant of PDMS is set to 4.44. By using the commercial electromagnetic simulation software HFSS, the dispersion curves of the spoof surface plasmons changing with different $h$ have been calculated and shown in Fig. 2(c). When the depth is increased from 6 mm to 8 mm, the curves depart away from the light line gradually. Hence, for the same wave vector $\beta$, the operating frequency becomes smaller, which suggests that the resonance frequencies could be tuned by changing the geometric parameters.
of the metal corrugated ring, as demonstrated in Fig. 2(d). The resonance peaks marked by $m_1$ - $m_4$ are corresponding to the dipole mode, the quadrupole mode, the hexapole mode, and the octupole mode, respectively. Since the quality factor ($Q$) of the hexapole mode $m_3$ is the highest of 578 (when $h = 6 \text{ mm}$), the $m_3$ mode would be used for sensing, where $Q$ is calculated by the formula $Q = f_0/f_{3\text{dB}}$, where $f_0$ is the resonance frequency of the resonator and $f_{3\text{dB}}$ is the 3dB bandwidth.

III. DESIGN OF THE MICROFLUIDIC CHANNEL
The schematic structure of the microfluidic channel is shown in Figs. 3(a)-3(c), which is etched in the PDMS substrate to simulate the blood vessel. Since the blood vessel is complicated, a complicated and central-symmetric microfluidic channel has been designed, which is concentric with the metallic corrugated ring and winding-shaped and interleaved with the top corrugated ring. The injected liquid is used to model the blood. To ensure the fluidity of the liquid under test, the center of the microfluidic channel is annular in order. The black arrows in Fig. 3(c) indicate the flow direction of the liquid. The detailed geometrical parameters of the microfluidic channel are listed in Table 1. In the simulation, the relative dielectric constant of DI water and 100% ethanol are set to be 79.5 and 9, respectively. The simulated reflection coefficients of $m_1$- $m_4$ modes have been shown in Fig. 3(d), when the sensor is filled with air, 100% ethanol and deionized (DI) water. It can be seen that compared with those of the $m_2$ and $m_4$ modes, the $Q$ factor of $m_3$ mode is higher and compared with that of the $m_1$ mode, the resonance frequency shift of $m_3$ mode is larger. Hence, the $m_3$ mode is used for sensing.

Next, a randomly shaped blood vessel is also investigated, as shown in Fig. 4(a), where the branch-shaped microfluidic channels have been designed, ensuring that the used fluid volume in the microfluidic channel is the same as that of the winding-shaped microfluidic channel shown in Fig. 3 (c). From the simulation results in Fig. 4(b), it can be seen that the resonance frequency changes from 5.353 GHz (air) to 5.298 GHz when the channel is filled with ethanol, and then it changes to 5.205 GHz when the channel is filled with DI water. Hence, the simulated resonance frequency shift is 148 MHz, which indicates the microwave plasmonic sensor is still sensitive for different kinds of blood vessels.

To validate the flexibility of the sensor, the flexible microwave plasmonic sensor with different bending angles ($\theta = 30^\circ$ and $60^\circ$) are simulated, as shown in Fig. 5(a). Figure 5(b) plots the simulated results when the microfluidic channel is empty or filled with polar solutions. It can be seen that the resonance frequency shift is still 160 MHz for $\theta = 60^\circ$, comparing to the planar sensor. The performances of the flexible plasmonic sensor with different bending angles are given in Table 2. When $\theta = 60^\circ$, the $Q$ factor at 5.5 GHz is 358, which is smaller than that of the planar sensor.

IV. EXPERIMENTAL VERIFICATIONS
The samples of PDMS microfluidic channel for the sensing application are fabricated from Juna Chip Technology.
TABLE 2. The performance of sensor with different bending angles.

| θ    | $f_{res}$ (GHz) | $f_{trans}$ (GHz) | $\Delta f$ (MHz) | Q   |
|------|-----------------|--------------------|------------------|-----|
| 30   | 5.627-5.562     | 6.472              | 155              | 325 |
| 60   | 5.507-5.428     | 6.347              | 160              | 358 |
| Planar | 5.232-5.310    | 6.392              | 160              | 578 |

FIGURE 5. (a) The 3D structure of microwave plasmonic sensor with the bending angle $\theta = 60^\circ$, (b) Simulated reflection coefficients of the flexible sensor when $\theta = 60^\circ$.

FIGURE 6. (a)-(c) The flexible PDMS microfluidic chip. (d) The microwave plasmonic sensor composed of printed metal corrugated ring on the ultrathin PET. (e) and (f) The flexible microwave plasmonic sensor sample.

Co. Ltd (Hefei, China). The microfluidic channel was made using soft lithographic techniques. The negative-tone UV photoresist SU-8 2100 (MicroChem Corp, Westborough, MA, USA) was spin-coated on 4” Si wafer with a thickness of 500 μm. Photolithography was performed to pattern microfluidic channel. The elastoplastic material PDMS (Sylgard 184, Dow Corning, Midland, MI, USA) mixture was degassed in a vacuum drying chamber and poured into the horizontally placed SU-8 molds. After curing and releasing the PDMS replica, PDMS replicas of 40 mm × 34 mm × 1.5 mm were cut. The PDMS replicas and 1.0 mm thick PDMS substrate were bonded by oxygen plasma in a vacuum drying chamber. The metal corrugated ring shown in Fig. 6(d) is printed on the PET membrane by use of inkjet printing technology. The conductive ink materials is Nano Ag inks, developed by Innovation (Beijing) Optoelectronics Co., Ltd, with the detailed specification of 50 nm size, 40% to 50% solid content, 1~2 mΩ square resistance. About 5 mL Ag inks are transferred to the ink storage tank and pipeline by use of the micro pipetting gun, and then sprayed on the PET membrane and kept at room temperature about 12 hours. The assembled microwave plasmonic sensor is shown in Figs. 6(e) and 6(f). It can be seen that the microwave plasmonic sensor can be flexible.

The final fabricated sample of the proposed microwave plasmonic sensor is illustrated in Fig. 7(a), where a 50 Ω connector is soldered at the input port. The reflection coefficients of the sensor are measured by using Agilent E5071C vector network analyzer (VNA), shown in Fig. 7(b). It can be seen that the simulation results agree well with the measurements. The experimental results of resonance modes marked by M1-M4 are located at 2.69 GHz, 4.44 GHz, 5.49 GHz, and 6.21 GHz, respectively. The corresponding simulated resonance frequencies are located at 2.39 GHz, 4.23 GHz, 5.35 GHz, and 6.05 GHz, respectively. The resonance frequency offset may be caused by the difference of the relative dielectric constant of the PDMS between the simulation and the sample. The measured Q value of the hexapole mode is 266.

The experimental setup for measuring the liquid solutions is displayed in Fig. 8(a), where the flexible and printed microwave plasmonic sensor is placed on top of the planar foam. To test the flexibility, the sensor with the bending angle 60° is attached to a cylindrical foam. The top view and the side view of the experimental platform using the...
bending sensor are shown in Figs. 8(b) and 8(c). The capillary tubes are connected to the inlet and the outlet of the PDMS microfluidic channel. The liquid solution is pushed into the channel through the syringe. The temperature is maintained at room temperature. The sensor is connected to the VNA to measure the changes of the reflected microwave signal. The measured reflection coefficients of the planar sensor for two different liquid solutions are plotted in Fig. 8(d). The measurement results agree well with the simulation results in Fig. 3(e). When there is no liquid in the channel, the measured resonance frequency is 5.303 GHz. When DI water is injected into the microfluidic channel, the resonance frequency obviously red-shifts. The resonance frequency is located at 5.156 GHz and the frequency offset can reach up to 147 MHz. When the 100% ethanol solution was injected, the resonance frequency is 5.243 GHz and the resonance frequency offset is 60 MHz. The measured results of the bending sensor are shown in Fig. 8(e), which agree well with the simulation results in Fig. 5(b). Comparing with those of the planar sensor, the resonance frequencies blue shift. When the channel is empty, the measured resonance frequency is 5.565 GHz. When DI water is injected into the microfluidic channel, the resonance frequency is 5.41 GHz. The resonant frequency offset of the bending sensor is 155 MHz.

The performances of different microwave microfluidic sensors are summarized in Table 3. It has been shown that the $Q$ factor would decrease when the materials under test is high-loss, such as polar solutions [31]. To evaluate the sensitivity of the microwave plasmonic sensors, the figure of merit (FOM) is defined by FOM = $Q \times \Delta f / \Delta \varepsilon$, where $Q$ is the quality factor, $\Delta f$ is the resonant frequency shift and $\Delta \varepsilon$ is the change of the relative dielectric constant [35]. Compared with other works in [28], [30], [33], [34], it indicates that the FOM is as high as 1178, which shows excellent sensitivity.

### V. CONCLUSION

The printed electrodes for detecting small DC electrical changes of the body have been fully investigated. Here a flexible and printed microwave plasmonic sensor for detecting liquid solutions is demonstrated. The sensor is composed of the printed spoof plasmonic resonator on the flexible PET substrate. The spoof plasmonic resonator is the metal corrugated ring fabricated by using inkjet printing technology. The simulated results agree well with the experimental measurements. It shows that the measured resonance frequency offset can reach up to 147 MHz when the DI water is injected into the microfluidic channel, which indicates that flexible microwave biological sensor is feasible. The FOM of the sensor reaches up to 1178. More works can be conducted for the practical applications of wearable and flexible microwave sensors for measuring subtle changes of blood glucose.

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