We present a flexible plastisol-based microfluidic process integrated with conductive nanoparticle composite polymer (C-NCP) electrodes for flexible active microfluidic devices on textile substrates. First, we characterize the stretchability and flexibility of both plastisol films and microfluidic channels. A maximum elongation increase of 37.5% is observed for plastisol films, and a maximum elongation increase of 17.5% is observed for microfluidic channels. We also demonstrate multiple levels of microfluidic channels. Using a new integrated fabrication process, a device that measures the conductivity of fluid between two electrodes is fabricated on a textile and successfully demonstrated and characterized. For this new fabrication process, flexible screen-printable Ag C-NCP, with resistivity of $2.12 \times 10^{-6} \Omega \cdot m$, is used for device electrodes. Commercial Ag epoxy, with resistivity of $1 \times 10^{-6} - 10 \times 10^{-6} \Omega \cdot m$, is also used to fabricate a second set of electrodes for comparison. The device is tested with saline solutions at different salt concentrations, and the current through each saline solution is measured at different voltages using both Ag C-NCP electrodes and Ag epoxy electrodes. The current increases linearly for a given voltage as the salt concentration increases, for devices with both Ag C-NCP electrodes and Ag epoxy electrodes.

© The Author(s) 2019. Published by ECS. This is an open access article distributed under the terms of the Creative Commons Attribution 4.0 License (CC BY, http://creativecommons.org/licenses/by/4.0/), which permits unrestricted reuse of the work in any medium, provided the original work is properly cited. [DOI: 10.1149/2.0191909jes]

Flexible and wearable microfluidic devices are among the newest wearable devices for applications in health monitoring, drug delivery systems, and bio-signal sensing. Flexible and wearable sensors that can conform to the body’s curved surfaces enable stable sensing and monitoring due to firm contact to skin without impeding motion. Many researchers are already developing wearable biofluid sensors such as perspiration sensors or lactate sensors. While these sensors are wearable in the sense that they can be worn on the body, there is a lack of development of devices that are integrated into clothing or that use textile substrates. The use of clothing for wearable sensors has many advantages. For example, the combination of bio-sensors with clothing can eliminate the need to use external devices that may have complicated attachment procedures. Also, lightweight wearable sensors on clothing enable real-time monitoring using low profile instrumentation without hindering human motion or activity. Textile-based microfluidic devices do exist, where the textile itself is used for the microfluidic “channels”. However, textile-based microfluidic devices suffer from many problems, such as difficulty in use for different types of liquids, and difficulty in repetitive use. Because the fabric channels absorb liquid, and biological components in the liquid may be attracted to textile fibers, resulting in loss of biological component or contamination. Thorough washing must be performed in order to remove contaminants. However, repeated washing may damage the devices. Other flexible devices exist that may be considered wearable. For example, paper-based flexible devices, some polydimethylsiloxane (PDMS)-based devices, and “tattoo” devices are developed by many groups. However, such devices suffer from limitations such as long fabrication times, requirement of clean room microfabrication facility for fabrication, and difficulty in repeat use.

To overcome limitations of other flexible and wearable microfluidics technologies, we have previously presented a new printing-based fabrication process (currently patent pending) for fully flexible, wearable and durable microfluidic channels and devices. For these microfluidic structures, a screen printing technique is used whereby plastisol ink (LFP-1060 crystal gel, QCM Polymers) is patterned on textile substrates. The technique employs paper as sacrificial substrates to form thin plastisol layers, with each layer of printed plastisol film patterned and bonded together to form microfluidic devices. The paper substrates are removed after bonding so that only multiple layers of plastisol film remain. The resulting microfluidic channels and devices are fully flexible, wearable, and durable, can be fabricated directly on textiles, and can be laundered.

We have previously presented single microfluidic channel layers and a simple passive mixing device in a publication focusing on the microfluidic channel fabrication process. We now present new work on integrating our plastisol-based microfluidic channels and structures with fully flexible conductive nanoparticle composite polymer (C-NCP) electrodes that are also based on plastisol polymer. The use of the same plastisol polymer for passive microfluidic structures and as a base polymer for C-NCP electrodes allows us to fabricate complex microfluidic devices and systems that are highly flexible and wearable, and that have active microfluidic and sensor components. Our group previously presented other polymer-based composite polymers and employed them in various applications such as tissue impedance measurement electrodes and microfluidic valves. We have also presented a conference paper employing plastisol-based C-NCP for electrocardiogram (ECG) electrodes for monitoring heart signals. We now present here for the first time multiple layers of plastisol-based microfluidic channels and plastisol-based C-NCPs that are integrated together for wearable microfluidic devices and systems on the same textile substrate.

In order to develop integrated devices and systems, the stretchability and flexibility of the plastisol film and plastisol-based microfluidic channels are first characterized. This new characterization is needed in order to facilitate development of more complex devices and systems, including multiple layers of channels and integration with electrodes and electronic routing. We also now present for the first time multiple layers of plastisol-based microfluidic channels on textiles for complex wearable microfluidic devices and systems. In addition, a microfluidic fluid conductivity sensor is developed to demonstrate the integrated fabrication process. The sensor consists of a microfluidic channel, a pair of electrodes (C-NCP electrodes or commercial conductive Ag epoxy electrodes), electrical routing for signal transfer, and contact pads for connection to external equipment for testing. Screen printable Ag C-NCP (previously characterized for a different application in a previously published conference paper) is used to fabricate...
electrodes, electronic routing, and contact pads. Lastly, the sensor device is successfully tested with saline solutions at different salt concentrations to demonstrate the example device.

**Stretchability and Flexibility Characterization of Plastisol Films and Microfluidic Channels**

**Stretchability of plastisol films and microfluidic channels.**—To test the stretchability of the plastisol films employed for microfluidic channels and devices, ninety samples of plastisol film are prepared. Each sample has the same width and length (30 mm width × 70 mm length), but has a different thickness. Samples are sorted into thirteen groups by thickness (65 to 100 μm, 100 to 135 μm, 135 to 170 μm, 170 to 205 μm, 205 to 240 μm, 240 to 275 μm, 275 to 310 μm, 310 to 345 μm, 345 to 380 μm, 380 to 415 μm, 415 to 450 μm, 450 to 485 μm, and 485 to 515 μm) with six to eight samples per group. In order to limit testing to only the plastisol films, each plastisol film is tested without a fabric substrate (not printed on fabric). Plastisol is printed on paper without a pattern, cured, and detached from the paper using methods similar to those reported previously. Each sample is stretched in increments of 200 μm, and the stretched length at which each sample fails is recorded.

To stretch each sample in increments of 200 μm, a micro-ruler with lines that are separated by 200 μm is prepared. Samples are attached to the ruler using adhesive tape such that only the middle of the sample (30 mm width × 40 mm length) is stretched. Figure 1 shows the ruler design and test scheme. The ruler is split into a right side and a left side with lines separated by 200 μm on both sides (see Figure 1a). A laser cutter is used to engrave the lines on the PMMA (Polymethyl methacrylate). Laser cutter settings of 3% of power (low power) and 90% of speed (fast speed) are used to pattern the lines with fine resolution. As shown in Figure 1b, one side (here, labelled as the “bottom”) of the sample is fixed so that only the other side (the “top”) can be stretched. By aligning the top line of the sample with the lines on the ruler, the sample can be stretched in increments of 200 μm, and the amount of total stretch (elongation increase) measured. The stretched sample is fixed for 30 seconds and stretched again manually (which takes about 5 seconds per stretch) without recovery.

As each sample is stretched, the film fails (breaks) at a certain elongation distance (breaking point). Figure 2 shows the breaking point
versus sample thickness for the different sample groups (thicknesses). The average film thickness for each group and the average breaking point for each group, in mm, are shown. The vertical error bars show the range of the breaking points and the horizontal bars show the range of thicknesses of each sample group. From these results, we observe that the breaking point increases as the thickness of the film increases until a certain point (around 390 μm), after which the breaking point tends to stay fairly constant (within 3%) regardless of the film thickness. For samples greater than 400 μm in thickness, the breaking point appears to level off, and compared to the initial (unstretched) length, a maximum of 37.5% of stretching (elongation) is observed before breakage. From these results, we may conclude that plastisol films less than 400 μm in thickness are optimal to maximize the stretchability of devices and to minimize the physical thickness and profile of fabricated devices.

In addition to the stretchability of plastisol films, the stretchability of microfluidic structures is also tested. In order to test the stretchability of microfluidic structures in plastisol, two simple microfluidic channels 500 μm in width, 15 mm in length, and approximately 100 μm in thickness, are prepared. As with the un-patterned plastisol samples, the channel devices are not patterned on fabric in order to test the stretchability of the devices alone. Figure 3 shows the design of the channels. One sample has a vertical channel (Figure 3a), and another sample has a horizontal channel (Figure 3b). For both channels, the bottom and top layers are designed to be approximately 150 μm in thickness, so that the total thickness of each device is designed to be 400 μm in order to maximize sample stretchability as previously discussed. The actual measured averaged thicknesses are around 426 μm for both samples. The variation is caused by the manual coating of the plastisol. The inlet and outlet each has a diameter of 2 mm. Each sample is stretched in increments of 500 μm until failure, and the width and length change of each channel are measured.

For the test, five additional distance rulers are prepared on a PMMA substrate. Figure 4 shows the design of the rulers. One ruler is designed for sample stretching similarly to that shown in Figure 1a for un-patterned samples, but the lines are separated by 500 μm instead of 200 μm. The other four rulers are designed to measure the dimension change of the channels. For these rulers, the length of the lines on the ruler is increased or decreased by 50 μm for adjacent ruler markings. Arrays of lines in different lengths are made as shown in Figure 4.

In order to measure change in each channel’s dimensions, the two microfluidic channels are attached and aligned on the PMMA micro-rulers, and stretched in increments of 500 μm, similarly to the method shown in Figure 1b. For each channel, the changes in length and width are measured. For example, to measure the increase in width of the horizontal channel, the channel is aligned to the left end line (500 μm) of the ruler shown in Figure 4e. Then as the channel width increases due to stretching, the line which contacts the channel’s edge is observed and the length of the line is recorded. Increased distance is calculated by subtracting 500 μm (initial width) from the length of the line which makes contact to the channel edge. Figure 5 shows photographs of the testing method for the horizontal channel and a microscopic image of the channel width measurement. Figure 5a shows the horizontal channel before stretching, and Figure 5b shows elongated channel under stretching prior to the breaking point. Figure 5c shows the channel at the breaking point. Figure 5d shows how the channel width is measured by the micro-ruler as described in Figure 4e. The initial width of the channel is 500 μm, and the width is increased upon stretching. The photograph shows that the channel width is stretched approximately 150 μm because the fourth line (650 μm) makes contact to the channel edge. The channel width change is measured in increments of 50 μm. For the horizontal channel, the length decreases...
and the width increases upon stretching. The maximum length decrease before breakage is 7.55% (from 19.2 mm to 17.75 mm) and the maximum width increase is 45.45% (from 0.55 mm to 0.8 mm). On the other hand, for the vertical channel, the length increases and the width decreases upon stretching. The maximum length increase is 15% (from 19.05 mm to 21.95 mm) and the maximum width decrease is 20% (from 0.5 mm to 0.4 mm). Compared to the stretchability of the un-patterned plastisol films, the microfluidic channel device could not be stretched as far before failure. Specifically, un-patterned 400 μm-thick plastisol films could be stretched approximately 15 mm (37.5%), but the channel devices could only be stretched approximately 7 mm (17.5%) before failure. This is likely because the channel device is composed of three separate plastisol films that are bonded with empty space in the middle to form the channel, creating a less robust structure. As Figure 5c shows, the damage caused by stretching occurs at the location of the channel.

Flexibility of microfluidic channels.—To demonstrate that the fabrication method is suitable for flexible and wearable devices and systems, the flexibility of microfluidic devices is also tested. While a similar but simpler qualitative test is performed on a microfluidic mixer fabricated on a textile substrate in a previous publication, the advantages of our fabrication method that we have not previously shown yet still result in a working device.

Multiple Layers of Wearable Microfluidic Channels

Previously we have demonstrated single layers of plastisol-based microfluidic channels on fabric substrates. However, one of the advantages of our fabrication method that we have not previously shown is that multi-layered structures can be easily fabricated by simply bonding more layers. Additional layers can be prepared with a laser cutter machine and bonded to form multi-layered structures. Figure 8 shows the design of an example microfluidic structure with multiple layers of microfluidic channels and interconnecting via holes. One sample has two channels in parallel on different levels (Figure 8a), and consists of a bottom layer, first channel layer, second channel layer, and top layer (Figure 8b). Each channel is designed to be 100 μm in width and height. A second sample has two channels that intersect at their centers, with the channels at right angles to each other (Figure 8c). Both channels are designed to have a height of 100 μm and a width of

![Figure 5](image_url)

**Figure 5.** Pictures of the sample and channel width measurement: (a) Horizontal channel before stretch. (b) Stretched channel. (c) Broken sample at breaking point. (d) Measurement of the increase in width of the horizontal channel. The initial channel width is 500 μm, and the width is increased upon stretching. It is shown that the width is increased by 150 μm.

![Figure 6](image_url)

**Figure 6.** Photographs of the paper pillars and the testing of microfluidic channel flexibility: (a) Paper pillars with different diameters (5 cm, 4 cm, 3 cm, 2 cm, 1 cm). (b) Test set-up. The flexible channel is attached on the paper pillar, and red ink is injected through the channel using a syringe pump at volumetric flow rate of 5 ml/min.

![Figure 7](image_url)

**Figure 7.** Working flexible microfluidic channel (plastisol only, no textile substrate): (a) Working channel wrapped twice around a paper pillar 1 cm in diameter. (b) Using ball-point pens to demonstrate flexibility that can be achieved yet still result in a working device.
Ag C-NCP, the epoxy is employed for one set of electrodes in order to compare the epoxy’s characteristics with those of the Ag C-NCP electrodes. This is done to compare the two materials, as Ag epoxy is a commonly used screen-printable material when conductivity is required. Figure 9c shows the design of the electrodes, electrical routing, and contact pads. The same Ag epoxy is used to form electrical routing and contact pads for the epoxy electrodes. However, Ag C-NCP is used to form electrical routing and contact pads for the Ag C-NCP electrodes. Each electrode (150 μm thickness × 800 μm width × 3 mm length) is inside the channel, thus making contact to the fluid. The electrical routing (150 μm thickness × 1.5 mm width × 4.5 mm length) is embedded in the device outside the channel for electrical isolation. Each contact pad (400 μm thickness × 2.5 mm width × 2.5 mm length) is open for connection to the digital multimeter. Thus, only the contact pads are exposed to air, with the electrodes and electrical routing all embedded in the device, either in the microfluidic channel (electrodes) or between polymer layers (routing). Fluid is injected into the channel by syringe pump, flows through the channel, and the current through the fluid is measured using the digital multimeter with a voltage applied between each set of electrodes (Ag epoxy or Ag C-NCP electrode pair).

Figure 10 outlines the fabrication process of the device. First, the bottom layer is screen printed on fabric, and C-NCP electrodes, electrical routing, and contact pads are screen printed on the bottom layer using Ag C-NCP (Figure 10a). We use 70 wt%-of Ag particles to prepare the Ag C-NCP in a similar manner to what we have reported previously. The adhesion between the bottom layer and C-NCP electrodes is very strong (as tested via tape test using techniques similar to other researchers), likely because the same plastisol ink forms the base polymer for each structure. Commercial Ag epoxy is next screen printed to pattern Ag epoxy electrodes, electrical routing, and contact pads (Figure 10b). The first channel layer is then patterned using a laser cutter and bonded to the bottom layer on the same level with the electrodes (Figure 10c). Next, contact pads for the two different types of electrodes are screen printed again and a second channel layer is prepared and bonded (Figure 10d). The second channel layer is designed to have both the channel and contact pad patterns so that electrical routing is covered by the second channel layer, thus embedding the electrical routing between the bottom layer and second layer. However, the contact pads are on the same level as the second channel layer because they are printed twice. Finally, contact pads are printed again, and the top layer is prepared and bonded to seal the channel (Figure 10e). The top layer is designed to have inlet, outlet, and contact pad openings so that the channel is sealed by the top layer, with only the contact pads exposed (not embedded between polymer layers in the device). As shown in Figure 10, the final device consists of four layers. The second channel layer is needed because the electrodes are on the same level with the first channel layer. Without the second channel layer, the channel would be blocked by the electrodes and injected fluid could not flow through the channel. Figure 11 shows a photograph of a fabricated device with close-up and microscopic images. Figure 11a shows the top view of the device and close-up of the active components (electrodes, electrical routing, and contact pads). Figure 11b shows a cross-sectional view of the Ag C-NCP contact pad, and Figure 11c shows the cross-sectional view of the channel on the Ag C-NCP electrode.

Experimental methods and results.—Saline solution samples with different concentrations of salt are employed to test the flexible microfluidic device. Polyethylene tubes with diameters of 0.86 mm (#427425, Intramedic) are connected to both the inlet and outlet of the device using a silicone sealant (Dow Corning 732) to allow liquid saline injection and waste removal during testing. Ten saline solutions are prepared at different salt concentrations (0.2 wt-%, 0.4 wt-%, 0.6 wt-%, 0.8 wt-%, 1.0 wt-%, 1.2 wt-%, 1.4 wt-%, 1.6 wt-%, 1.8 wt-%, and 2 wt-%). An alternating current (AC) voltage at 10 kHz is applied between each type of electrodes separately to allow for comparison. For each saline concentration, the applied AC voltage is varied from 0.3 V peak-to-peak to 3V peak-to-peak by 0.3 V peak-to-peak...
Figure 9. Flexible and wearable fluid conductivity sensor design: (a) Cross-sectional view of the device showing one electrode, its electrical routing, and contact pad. The electrode is inside the channel, and the electrical routing is embedded in the device. (b) Top-down view of the device. A pair of Ag C-NCP electrodes and a pair of Ag epoxy electrodes are aligned with a microfluidic channel. (c) Detail of the electrodes’ design. Each electrode is connected to the contact pad through the electrical routing.

Figure 10. Integrated fabrication process for flexible and wearable conductivity sensor: (a) Print the bottom layer on fabric, and print Ag C-NCP electrodes, electrical routing, and contact pads on the bottom layer. (b) Print Ag epoxy electrodes, electrical routings, and contact pads using Ag epoxy ink. (c) Prepare the first channel layer, align the channel to the electrodes, and bond to the bottom layer on the same level with electrodes. (d) Print contact pads for both types of electrodes, and prepare and bond the second channel layer. Contact pads are on the same level with the second channel layer. (e) Print the contact pads again, and prepare and bond the top layer. Only the contact pads and inlet/outlet holes are exposed on the device, with all other structures (electrodes, channel, etc.) embedded in the device.
Figure 11. Photographs of the device with close-ups and microscope images: (a) Top view of the device and close-up of the conductive active component. The close-up shows a Ag C-NCP electrode in the microfluidic channel so that fluids can flow on the electrode. (b) Cross-sectional view of the Ag C-NCP contact pad. (c) Cross-sectional view of the channel on the Ag C-NCP electrode.

saline solution concentration the same. Next, the same measurement is repeated at different salt concentrations, but the same applied voltage, to check the correlation between salt concentrations and measured current. Ag epoxy electrodes are then tested with the same procedure. The current measurements are performed three times for all saline concentrations at each applied voltage, and the channel is washed between measurements of each saline concentration to minimize the effect of fouling for the proof-of-concept devices. From these measurements, the average current is calculated for both Ag C-NCP electrodes and Ag epoxy electrodes. Figure 13a shows the results from the Ag C-NCP electrode pair, and Figure 13b shows the results from the Ag epoxy electrode pair.

For the saline solutions, filtered water (control water) is used, but a small current is measured due to a small amount of minerals in the control water. When de-ionized water is used, no current is measured as shown in Figure 13. For both cases, the current appears to increase linearly with the applied voltage, and the variations for each measurement are very small. The standard deviation is smaller than 0.3%. The error bars are shown in the Figure 13, but they are not visible at the scale of the graphs. In addition, the current increases as the salt concentration increase, as expected due to the higher conductivity with increasing sodium ion concentration. As the salt concentration increases in increments of 0.2 wt-%, the current increases approximately by 1.17 times for the Ag C-NCP electrodes and by 1.22 times for the Ag epoxy electrodes. Furthermore, for a salt concentration increase from 0.2 wt-% to 2.0 wt-%, the current increases approximately 3.98 times for the Ag C-NCP electrodes and approximately 5.57 times for the Ag epoxy electrodes. From these results, we may conclude that both types of electrodes function properly for current measurements across all concentrations of saline solution tested, but the Ag epoxy electrodes’ sensitivity to the sodium ions’ concentration change is higher than that of the Ag C-NCP electrodes.

Figure 14 shows a comparison between Ag C-NCP electrodes and Ag epoxy electrodes. The current change at a fixed salt concentration and the current change at a fixed voltage are compared. When the salt concentration is same, the current increases linearly as the applied voltage increases for both types of electrodes (Figure 14a). However, the measured current from the Ag epoxy electrodes is 1.4 times higher than the current from the Ag C-NCP electrodes. The error bars are shown in Figure 14, but are small compared to the scale of the graphs. Over the three measurements taken, maximum deviations of $-1.59\%$ and $+3.17\%$ are measured for the Ag epoxy electrodes, and maximum deviations of $-3.12\%$ and $+2.94\%$ are measured for the Ag C-NCP electrodes. The current also increases linearly as the salt concentration increases when a fixed voltage is employed (Figure 14b) for both types of electrodes. However, the current from the Ag C-NCP electrodes is
Figure 13. Current measurement results: (a) Results from Ag C-NCP electrodes. As expected, the current appears to increase linearly as the applied voltage increases. In addition, also as expected, the current increases as the concentration of salt increases. (b) Results from Ag epoxy electrodes. The current increases linearly as the concentration of salt increases and applied voltage increases.

smaller than the current from the Ag epoxy electrodes. Over the three measurements taken, maximum deviations of $-0.33\%$ and $+0.21\%$ are measured for the Ag epoxy electrodes, and maximum deviations of $-1.09\%$ and $+1.89\%$ are measured for the Ag C-NCP electrodes. In all cases, the currents from the Ag C-NCP electrodes are smaller than those of Ag epoxy electrodes. This is because Ag C-NCP is less conductive than Ag epoxy. From the tests in Figure 13 and Figure 14, it is observed that the conductivity and sensitivity of Ag C-NCP electrodes are lower compared with the Ag epoxy electrodes. However, the Ag C-NCP electrodes function in a similar manner, and are much more flexible than Ag epoxy electrodes, which is beneficial to wearable and flexible systems.

Figure 14. Comparison between Ag C-NCP electrodes and Ag epoxy electrodes: (a) Results at 2.0 wt-% of salt concentration with change of applied voltage. (b) Results at 3V of voltage with change of salt concentration. Though the conductivity and sensitivity of Ag C-NCP are lower than those of Ag epoxy, Ag C-NCP electrodes still work in a similar manner with Ag epoxy electrodes.
In this paper, a fabrication process for flexible and wearable plastisol-based microfluidic devices is presented and characterized. With the characterization, complex devices including multiple layers of channels are developed. Flexible and wearable microfluidic devices are also integrated with Ag conductive nanoparticle composite polymer (C-NCP) electrodes for flexible active microfluidic devices and systems on textile substrates. A device that measures the conductivity (current versus voltage) of fluid between two electrodes is fabricated, and used to measure current between two different types of electrodes at different voltages and saline solution concentrations.

Stretchability tests show that the plastisol films and microfluidic channels are stretchable; plastisol films with different thicknesses are stretched until failure, and the maximum stretched distance is recorded. The degree to which samples can be stretched before failure increases with increased thickness until a certain thickness after which the stretched length appears to level off regardless of the thickness of the plastisol films. A maximum stretch of 37.5% is observed for plastisol films. In addition to the stretchability of plastisol films, the stretchability of microfluidic structures is also tested. Two simple microfluidic channels are prepared, and the same stretchability tests are performed. Compared to the stretchability of the un-patterned plastisol films, the microfluidic channel device could not be stretched as far before failure. A maximum of 17.5% of stretch is observed for the channels before failure.

A fully flexible microfluidic device that measures the conductivity through the fluid between two electrodes under an applied voltage is also fabricated on fabric with flexible Ag C-NCP. For this sensor, Ag C-NCP is prepared using the same plastisol ink as the microfluidic channels. The Ag C-NCP shows stable electrical resistivity of $2.12 \times 10^{-6}$ $\Omega \cdot m$ at 70 wt-% of Ag particles. Commercial Ag epoxy is also used to form two other less flexible, but lower resistivity ($1 \times 10^{-6}$ to $10 \times 10^{-6}$ $\Omega \cdot m$) electrodes for comparison. Each set of electrodes is inside the channel so that fluid flows over the electrodes. Electrical routing is embedded in the device, and is employed to transfer electrical signals to the contact pads. Contact pads are exposed for connection to external equipment for testing. For the tests, saline solution samples with different concentrations of salt are employed to test the flexible microfluidic device, and an alternating current (AC) voltage is applied between each type of electrodes separately to allow for comparison. Both Ag C-NCP electrodes and Ag epoxy electrodes show a linear increase in current as applied voltage increases, and also as salt concentration increases, but the conductivity and sensitivity (slope) of the current-voltage relationship using the Ag C-NCP electrodes are lower compared with those of the electrodes fabricated in Ag epoxy. However, the results for both types of electrodes are qualitatively similar, with Ag C-NCP electrodes having the advantage of greater flexibility.

With the introduced fabrication method, limitations found in previously developed wearable microfluidic sensors, such as inflexibility, complicated fabrication method and need for cleanroom facility, and long-time fabrication times, are addressed. The new fabrication method has the advantages of inexpensive materials and equipment, short time processing, and very simple fabricated method. In addition, fully flexible microfluidic devices can be fabricated on fabric, and flexible electronic components also can be combined in the device using C-NCPs and the new integrated fabrication process.

Acknowledgments

This project is supported by Dr. Gray’s NSERC (The Natural Sciences and Engineering Research Council of Canada) Discovery Grant. Simon Fraser University and the authors have applied for a patent on the basic fabrication process. All devices are fabricated in the Simon Fraser University Microinstrumentation Lab and ENSC student fabrication (laser cutter) facilities, with profilometry performed in the SFU ENSC microfabrication facility. The authors would also like to thank current and former members of the Microinstrumentation Lab for valuable discussion.

ORCID

D. Chung https://orcid.org/0000-0002-7540-0426

References

1. H. Jung, J. Moon, D. Back, J. Lee, Y. Choi, J. Hong, and S. Lee, *IEEE Trans. Biomed. Eng.*, 59, 5, 1472 (2012).
2. W. Huang, S. Hu, K. Liu, S. Chen, and D. Liu, *J. Control. Release*, 139(5), 221 (2009).
3. G. Matreuz, L. Florea, and D. diamond, *Sens Actuators B Chem.*, 211, 403 (2015).
4. W. Gao, S. Enaminedjad, H. Y. Nyeim, S. Challa, K. Chen, A. Peck, H. M. Fahad, H. Ota, H. Shiraki, D. Kiriya, D. Lien, G. A. Brooks, R. W. Davis, and A. Javey, *Nature*, 529, 509 (2016).
5. W. Jia, A. J. Bandodkar, G. Valdés-Ramírez, J. R. Windmiller, Z. Yang, J. Ramírez, G. Chan, and J. Wang, *Anal. Chem.*, 85(14), 6553 (2013).
6. G. Liu, C. Ho, N. Slapper, Z. Zhou, S. E. Snelgrove, M. Brown, A. Grabinski, X. Guo,Y. Chen, K. Miller, J. Edwards, and T. Kaya, *Sens Actuators B Chem.*, 227, 35 (2016).
7. A. Nilghaz, D. H. B. Wicaksono, D. Gustiono, F. A. A. Majid, E. Supriyatno, and M. R. A. Kadir, *Lab Chip*, 10, 477 (2010).
8. J. Moon, D. Back, Y. Choi, K. Lee, H. Kim, and S. Lee, *J. Micromech. Microeng.*, 20, 025032 (2010).
9. M. S. Fernandes, K. S. Lee, R. J. Ram, J. H. Correia, and P. M. Mendes, 32nd Annual International Conference of the IEEE EMBS, 5503 (2010).
10. A. J. Bandodkar, W. Jia, and J. Wang, *Electrochem.*, 27, 562 (2015).
11. D. Chung and B. L. Gray, US2017012808A1 (2017).
12. D. Chung and B. L. Gray, *J. Micromech. Microeng.*, 27, 11, 115009 (2017).
13. D. Chung, A. Khosla, B. L. Gray, A. M. Parameswaran, R. Ramaseshan, and K. Kohli, *J. Electrochem. Soc.*, 161(2), B3071 (2014).
14. A. Khosla, B. L. Gray, *Microfluidics, BioMEMS, and Medical Microsystems IX*, 2011.
15. D. Chung, A. Khosla, and B. L. Gray, *Proc. SPIE* 9060, 227 (2015).
16. C. Peng, Z. Chen, and M. K. Tiwari, * Nat. Mater.*, 17, 355 (2018).