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A Soft Stretchable Sensor: Towards Peripheral Nerve Signal Sensing

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ABSTRACT

We propose a 3D-printable soft, stretchable, and transparent hydrogel-elastomer device that is able to detect simulated 'nerve' signals. The signal is passed to a conductive hydrogel electrode through a non-contact method of capacitive coupling through polydimethylsiloxane (PDMS). We demonstrate that the device is able to detect sinusoidal waveforms passed through a simulated 'nerve' made from conductive hydrogel over a range of frequencies (1 kHz – 1 MHz). Analysis of signal detection showed a correlation to the electrode contact area and a V_{in}/V_{out} *of larger than 10%. This provides the framework for the future development of a soft, 3D-printable, capacitive coupling device that can be used as a cuff electrode for detecting peripheral nerve signals.*

INTRODUCTION

Amputees rely on prosthetic devices to serve as extensions of their body to perform tasks on a day-to-day basis. The ideal prosthetic for completing many activities would be lightweight, lifelike, affordable, and -- above all -- fully functional [1]. A sophisticated method of accurately controlling a prosthetic device would involve the use of neuronal signals to convey the user's intentions. The detection of neuronal signals from biological organisms has remained an area of importance for decades, and numerous methods for detecting neuronal signals from the central nervous system (CNS) and peripheral nervous system (PNS) have been reported in the past [2]. Many of these previously reported devices rely on the use of expensive materials such as gold or platinum. Additionally, they often result in devices that are not mechanically compatible with biological tissue and could harm organisms over long periods of time [3,4].

Recent advances in 3D-printing and the development of novel soft materials have the potential to create a neural interface that could function without the aforementioned issues. In previously reported work, dielectric materials such as polydimethylsiloxane (PDMS) and ionically conductive poly(acrylamide) (PAAm) hydrogels have been used to create soft stretchable devices that can be printed to sub-millimeter resolution and are able to transmit electrical signals over long distances [5-7]. In this study, we use these materials to create a device that can sense electrical signals in a non-contact manner via capacitive coupling. We demonstrate the capability of this device to detect signals over a large range of amplitudes and frequencies, which makes this a potential technology for applications such as PNS signal detection.

EXPERIMENTAL

PDMS and curing accelerator (Shin-Etsu Silicones) were combined in a 10:1 ratio and mixed for 2 min in a planetary centrifugal mixer (Thinky, ThinkyMixer ARE-300) at 2000 rpm. The PDMS precursor was extruded by hand onto a pre-cut 5 mm acrylic sheet using a 5 mL syringe and cured for approximately 12 hours in an oven at 65 °C. The PDMS was then treated with O_2 plasma (SPI Supplies, Plasma Prep II) at an O_2 pressure of 982 mTorr, vacuum pressure of 275 mTorr, and radio frequency power of 80 W for 60 s. After treatment, the PDMS was submerged in a sealed container filled with DI H_2O (resistivity = 18.2 MΩ cm) for storage to preserve the hydrophilicity.

The LiCl doped hydrogel precursor to be extruded onto the PDMS surface was created as per previously reported methods [5]. Prior to extrusion, the PDMS was removed from the DI and blown dry with N_2 gas. The hydrogel precursor was then deposited onto the PDMS by extrusion through a syringe. After deposition, the precursor was exposed to UV light (8 W 365 nm, UVP, UVLS-28 EL, effective dose of ~6 mW cm^{-2}) until fully cured in a N_2 filled humidity controlled environment (43%), which was maintained by blowing N_2 gas through a saturated solution of potassium carbonate. Afterwards, the sample was covered in another layer of PDMS by extrusion through a syringe and cured by baking for approximately 12 hours in an oven at 65 °C. The sample was removed from the oven and left to soak in DI water for at least 12 hours (until testing) to rehydrate the hydrogel.

To simulate peripheral nerve signal detection, a sample `nerve' made of the same conductive hydrogel described above was deposited onto the device using a syringe. It should be noted that all materials used to fabricate the device are compatible with 3D-printing methods, even though in this work they were dispensed by hand using a syringe [5]. The simulated nerve was suspended vertically with an insulated gripper and wires were connected from a function generator (Keysight Technologies, Model 33500B) to the `nerve' to pass a sinusoidal signal with varying peak-to-peak amplitude (.1 – 1 V) and frequency (1 kHz – 10 MHz). Wires were brought in contact with the hydrogel encapsulated in the PDMS by piercing the PDMS and were connected to an oscilloscope (Keysight Technologies, Model DSO1004A). All leads to ground were grounded together and a schematic of the final experimental setup is shown in Figure 1. A total of 3 samples were made and the signal output from each of the printed hydrogel sensors were recorded from the oscilloscope relative to the input signal. The recorded signal was analyzed in relation to the contact area between the `nerve' and the PDMS above the electrode, which was measured in ImageJ.

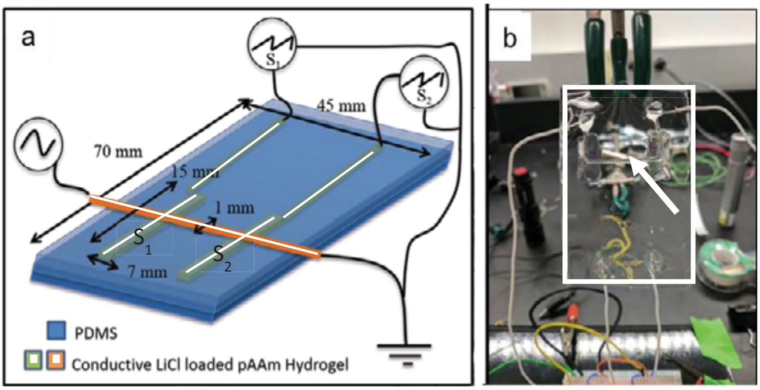

Figure 1: a) Schematic for passing simulated nerve signals through a model `nerve' made of conductive LiCl doped pAAm hydrogel (orange w. white stripe) through PDMS (blue) to a soft electrode made from the same LiCl loaded pAAm hydrogel (green w. white stripe) via capacitive coupling. S_1 and S_2 indicate the output from each electrode pad displayed on an oscilloscope. b) Photograph of a typical hydrogel-elastomer device (vertically suspended, white rectangle) in the experimental setup with the 'nerve' indicated by the white arrow.

DISCUSSION

Considerations of Creating the Nerve Sensor

The proposed device uses capacitive coupling to detect a signal in a non-contact method through a dielectric material (PDMS). In this specific case, a signal passes through the `nerve,' which is in contact with the sensor. The signal passes through the sensor dielectric and is transmitted through the conductive hydrogel wires to the oscilloscope. A nerve signal was simulated by passing a sinusoidal signal through a line of conductive hydrogel (approx. 1 mm track width) using a function generator after taking into account the following considerations. Compound action potentials (CAPs) of sensory nerve signals propagate down the axon at speeds of up to 120 m/s, but peripheral nerve signals travel at much slower speeds of about 50 - 60 m/s [8,9]. In a previously reported cuff electrode implanted inside an animal on a peripheral nerve, the two electrodes were spaced at an approximate distance of 2 mm apart [10]. A conservative estimate could be made to consider a CAP passing down the axon of a nerve at 95 m/s. This signal would reach the second electrode approximately 0.2 ms after the first.

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Therefore, for the proposed system to be able to distinguish a peripheral nerve signal, it would need to resolve signals of at least 50 kHz. Additionally, it must detect signals of a comparable amplitude to the CAP of a peripheral nerve, so attempts to find the smallest detectable signal were made.

Quantifying signal detection

A range of frequencies and amplitudes of a sinusoidal waveform were generated and passed through the device. Initially, an amplitude of 1 V_{pp}, where V_{pp} indicates the peak-to-peak voltage, was passed through the hydrogel `nerve' at a range of frequencies to determine the highest possible frequency that could be resolved and to determine the effects of frequency on signal detection. The coupling was reported as the signal detection efficiency, (% V_{out} /V_{in}), where V_{out} is the peak-to-peak output voltage and V_{in} is the peak-to-peak input voltage. This process was repeated with the signal passing in the opposite direction of the conductive gel to determine if there was any difference in the magnitude of coupling between the two electrodes. The results for one of the samples is shown in Figure 2a. The coupling is reported for each electrode pad (S_1 and S_2) relative to the initial input voltage as measured from the function generator when the signal was not passed through the device. The default setup has the input source closest to S_1 and then closest to S_2 when `flipped.'

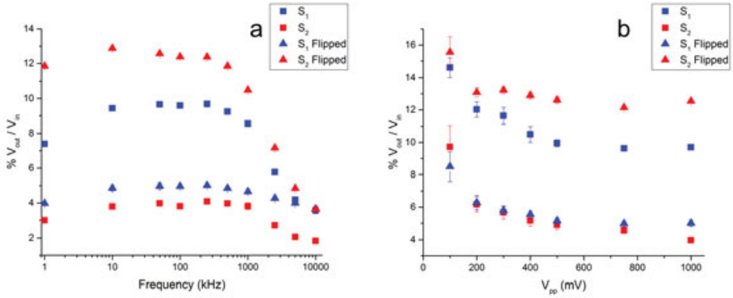

Figure 2: a) The measured coupling signal detection efficiency for a sinusoidal waveform over a range of different frequencies. b) The measured signal detection efficiency (% V_{out} / V_{in}) for a sinusoidal waveform over a range of amplitudes. Each electrode pad is denoted S_1 (blue) or S_2 (red) with either the input signal closest to S_1 (squares) or closest to S_2 (triangles).

Detection efficiency slowly increased with frequency from 1 kHz to 10 kHz and remained at a maximum of about 10-12% of the input signal until 1 MHz. At frequencies past this, the detection efficiency dropped off until it was indiscernible from noise (~12 mV_{pp}) at around 10 MHz. When swapping the side of the signal source, each electrode theoretically gave the same signal detection efficiency when the input source was closest to it. However, when the input was changed, it was observed that different electrodes of certain samples showed better coupling than others. After measuring variables such as distance to input source or PDMS thickness and finding no significant difference between the two sides, the variability was explained by a relation of the detection efficiency to the contact surface area between the PDMS and `nerve' model. The contact area of each electrode was estimated using ImageJ and the results comparing

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the contact area to the coupling efficiency for a single frequency (50 kHz) in two of the samples are shown in Table 1.

Table 1: A comparison of the measured contact areas of the conductive hydrogel `nerve' on the electrode pad to the signal detection efficiency when the input power source is closest to S_1 (default) or when the input power source is closest to S_2 (flipped). S_1 and S_2 refer to the output from each electrode as shown in Figure 1.

	Contact area (mm^2)	Default (% V_{out}/V_{in})	Flipped (% V_{out}/V_{in})
Sample 1, S1	7.8 ± 0.3	9.7 ± 0.1	5.0 ± 0.2
Sample 1, S2	9.2 ± 0.1	4.0 ± 0.2	12.6 ± 0.1
Sample 2, S1	6.8 ± 0.2	9.8 ± 0.1	4.3 ± 0.2
Sample 2, S2	4.5 ± 0.2	3.5 ± 0.1	8.1 ± 0.1

It was observed that electrodes with a larger contact area (Sample 1, S_2 and Sample 2, S_1) provided better coupling of the signal. For example, in Sample 1, S_2 had a larger contact area. When the source was closest to S_1 the detection efficiency was 9.7%, but when the input source was closest to S_2, the detection efficiency rose to 12.6%. These results indicate that the proposed device is able to resolve signals over a large range of frequencies with a conservatively reported value of 10% efficiency, and that this detection efficiency shows a correlation to the contact surface area between the device and the signal source. It is anticipated that the signal detection efficiency may in fact be greater than the reported values since the amplitude of the input signal, V_{in}, was measured at the signal source and not at the location of the the electrode pad. Therefore, the signal amplitude at the point of contact should be lower than that at the waveform generator because the whole line of conductive hydrogel should theoretically act as a voltage divider. Therefore, using the amplitude at the point of contact should lower the input signal magnitude, increase the detection efficiency, and give a more accurate representation of what the device is capable of. This also explains why the coupling values of the further electrodes are significantly smaller than those recorded at the closer electrode.

After establishing the range of frequencies that a signal could be detected over, attempts to determine the lower boundary of detectable amplitudes were made. Sinusoidal waveforms of 50 kHz with amplitudes ranging from V_{pp} = 100 mV to 1 V were passed through the device and the detected signals were recorded on an oscilloscope (Figure 2b). A frequency of 50 kHz was chosen based on the speed at which action potentials travel in peripheral nerves as discussed previously.

CONCLUSIONS

Devices used to sense PNS signals generally involve the use of expensive materials that may be mechanically incompatible with biological tissue over prolonged periods of time. A hydrogel-elastomer sensor made from soft, stretchable, 3D-printable materials and its ability to detect simulated nerve signals through capacitive coupling is explored. It was observed that this device was able to detect sinusoidal waveforms over a range of frequencies (1 kHz - 1 MHz) and amplitudes (V_{pp} = .1 V - 1 V). Theoretically, signals at much lower amplitudes could be detected by the device as well, however, this could not be demonstrated in the current experimental setup. Although the simulated 'nerve' signals detected were not action potentials, the frequencies resolved by the device indicate that it should theoretically be able to detect CAPs from the PNS. The detection efficiency, (% $V_{out}/V_{in)}$, was greater than 10% and was correlated to the contact area between the `nerve' and the PDMS above the electrode pad.

Overall, the work presented here shows the potential of using these materials in the creation of a soft, stretchable, and transparent electrode capable of peripheral nerve signal sensing via capacitive coupling. Eventually, this could lead to the creation of a low cost and mechanically compliant neural interface that could be used in applications such as neuroprosthetic devices

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