STRETCHABLE TACTILE SENSING SKIN FOR INTELLIGENT PROSTHETIC LIMBS

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ABSTRACT

In this work, we report for a first time a stretchable tactile sensing skin that senses distributed contact force and is capable of conforming to curved surfaces, such as those of prosthetic limbs. The sensor comprises of three elements – substrates, force transduction elements and electric leads. All these elements are stretchable. The 2D force sensing array consists of 14 nodes spreading over an area of $4\text{cm} \times 7\text{cm}$. We demonstrate here that the sensor skin can endure up to 8% of tensile strain repeatedly while the metal conductors are fully functioning, and with a detection limit of 1.57 mN normal force. The initial feasibility study shows promise for integrating such skin with a prosthetic limb part.

INTRODUCTION

Tactile sensor is defined as a device or system that can measure a given property of an object or contact event through physical contact between the sensor and the object [7]. The measured properties can include shapes, texture, temperatures, hardness, as well as normal contact force. Tactile sensing plays a key role in the fields where physical manipulation and tactile exploration are involved, such as surgeries [8] [9] [10], robotics [11], and entertainment, etc.

Tactile sensors integrated with prosthetic limbs that measure normal contact promise to restore the touch sensation to the disabled, allowing enhanced dexterity and manipulation capabilities. One premise for such application is that the sensors should be able to conform to the 3D-flexed surfaces while functioning. This criterion not only requires stretchable materials or structures for the carrier skin, but all transducers and electrical conductors to be made stretchable. Existing MEMS tactile sensors, mostly silicon based, do not meet such need.

One conceptual architecture was proposed by Wagner, et. al. [13], that an electronic tactile skin ideally should incorporate subunits of tactile sensors and actuators arranged in array configuration and deformable conductive leads running in lines and columns inside a stretchable continuous substrate. Here we adopted a similar configuration where each unit contains one sensing node and two stretchable electric leads. Adjacent leads are shared in order to minimize wiring complexity.

The stretchable sensing skin in this work consists of three stretchable components – the polydimethylsiloxane (PDMS) thin film substrate, the nanocomposite force sensing resistors (FSR) made of carbon nanofiber suspension in PDMS, and the electric conductors made of micro fluid channels filled with liquid metal alloy. Figures 1(a) – (f) are photos of the sensing skin undergoing various deformation modes. Figure 1(g) is a photo of the sensing skin conforming to the curved surface of a limb part. Figure 1(h) illustrates the conceptual design and function of the sensing skin. Figure 1(i) shows representative experimentally recorded time traces when such skin and a commercial load cell are subject to the same normal contact force – impulses in this case. The lower trace is the output from the load cell and

the upper trace is that from the sensing skin, demonstrating the skin's ability of capturing contact events in real time.

Figure 1: Conceptual design and function of the sensing skin. (a)-(f) Sensing skin under various deformation modes. (g) Sensing skin conformal to the curved end of a prosthetic limb part. (h) Conceptual illustration of design and function of sensing skin. (i) Signal comparison between sensing skin (upper) and a commercial load cell (lower) when subject to the same contact event

In this paper, the important characteristics and of single stretchable elements – FSRs and conductors are investigated separately, followed by discussion of the fabrication process flow, and concluded by the characterization results of the device as a whole.

MATERIAL OPTIONS AND DESIGN

Nanocomposite FSRs as embedded force sensors

It has been reported that force sensing resistors (FSR) can be made by doping polymers with conductive or semi-conductive particles or fibers, like carbon nanotubes or carbon nanofibers. The resistivity of the composite is extremely sensitive to the change of volume fraction of the nanofiller at the so-called percolation region. An external strain can change the volume fraction due to the difference in the stiffness between the filler and matrix materials thus can change the apparent resistivity of the mixture. This is the basic underlying mechanism for FSR's resistive strain sensitivity.

Strain-sensitive nanocomposite made of carbon nanofibers dispersed in polydimethylsiloxane (PDMS) matrix is used as force sensing resistors (FSR) in this work. It is chosen as embedded force sensors for its compatibility with the PDMS substrate. The FSR's resistive sensitivity to normal contact force has been characterized and utilized in our previous work [1].

In this work, we demonstrate that FSR's baseline

resistivity follows a linear relationship with the weight concentration of the nanofillers of the mixture on the semi-log plane, as shown in Figure 2. With this chart, the mixing ratio can be tuned in order to achieve desired resistance of the FSRs. In this work, 10% weight ratio is chosen for an ultimate resistivity of 1000 $\Omega\cdot\text{m}$.

Figure 2: FSR base-line resistivity as a function of wt.% concentration of carbon nanofibers on a semi-log plane

Micro liquid metal channel as conductors

Various approaches to make stretchable conductors on polymeric substrates have been investigated by researchers [2] [3] [4]. However, due to debonding failure and large resistivity, none have yet achieved ideal and robust conductors that endure repeating tensile loading on stretchable substrate.

Our group has pioneered the use of liquid metal alloy as conductors. In this work, it is experimentally proved that micro fluid channels filled with liquid metal alloy are ideal conductors for the stretchable sensors.

Test samples with liquid metal filled channels are prepared and four different widths are explored (See Figure 4). The liquid metal alloy used here is formed by using 3 metals – gallium, indium and tin. The melting point can range from room temperature to as low as -20°C , depending on the mixing ratio of the compound. Some additional advantages of such liquid metal alloys include good conductivity, wettability (due to gallium oxide formation), and non-toxicity [6].

Figure 3 shows the I-V characteristics of the liquid metal conductors of various widths. The straight lines on the I-V plane are strong indicators that the liquid metal channels are ideal Ohmic conductors. Figure 4 shows the measured resistance increases with applied tensile strain in a linear fashion. The conductors function repetitively when undergoing up to 8% tensile strain with a maximum resistance of 6.2 Ω .

We have also demonstrated in our previous work [5] that by introducing liquid metal as conductive enhancement, the contact resistance between nanocomposite FSRs and metal electrodes is dramatically reduced. This is another reason why liquid metal conductors are preferred.

FABRICATION

The fabrication involves two parallel processes followed by an assembly step, as is illustrated in Figure 5 and Figure 6. The FSR layer (Figure 5) and the micro channel layer (Figure

6) are prepared separately, and then bonded together (Figure 6).

Figure 3: I-V characteristics of liquid metal conductors of difference widths – 50, 100, 175 and 224 microns.

Figure 4: A plot of the resistance values of liquid metal leads as a function of applied longitudinal strain, with photo of the test sample.

The patterning of FSR starts with photolithography on a glass wafer. After exposure of the photoresist, nanocomposites are applied. A soft blade is used to squeegee the excess of nanocomposites, leaving only those inside the recess regions created by the photoresist. After baking at 90°C in the convection oven for hardening, the photoresist is dissolved in acetone. A thin layer of PDMS is either poured over or spin-coated on top of the FSRs, and the resulted thickness of PDMS coating can be controlled by the volume or the rate of spinning. After baking, PDMS and FSRs will bond firmly due to covalent bonding, thus can be peeled off from the glass substrate easily.

The patterning of micro channels starts with photolithography on a glass wafer, where the height of the channels is defined by the thickness of the photoresist. An

annealing step involving maintaining the samples at an elevated temperature is introduced. After annealing, the sharp corners of the channels are rounded and flow resistance during the injection process of the liquid metal can be reduced. PDMS is either poured over or spin-coated on top of the channels to form a thin film, the thickness of which can be controlled by volume or the rate of spinning. Injection holes are punched after the PDMS film is released from the glass substrate.

Figure 5: The process flow for patterned FSR layer. (a) Pattern photoresist to define the geometries of FSRs. (b) Apply carbon nanofiber / PDMS composites to the recessed regions (c) Squeegee with a soft blade to remove excess composites and shape FSR. (d) Bake FSR and remove photoresist (e) Pour or spin-on PDMS to a desired thickness. (f) Bake and release.

The FSR layer and the channel layer are bonded at 90°C in the convection oven after oxygen plasma treatment. Afterwards, liquid metal alloy is injected at a pressure of 20 psi using a home-build injection pump. The injection holes can be sealed with epoxy after external wiring is completed.

Figure 6: The process flow for micro channel layer and assembly. (a) Pattern photoresist to define the geometry of the

micro channels. (b) Anneal at 115 °C for photoresist reflow. (c) Pour or spin-on PDMS to a desired thickness. (d) Peel PDMS from the glass substrate. (e) Punch holes for liquid metal injection. (f) Bond FSR layer and channel layer after oxygen plasma treatment. (g) Inject liquid metal and seal with epoxy.

Figure 7: Resistance changes of the device as a function of the magnitude of the normal loading stress

DEVICE CHARACTERIZATION

The sensor's performance is characterized on a home-built automated system. A 1KG commercial load cell is used to monitor the applied force while a motorized stage is driving it at the velocity of 1mm/s towards the sensing skin. Two sets of data from both the reading of load cell and the output of the sensing skin are collected at a sampling rate of 200 Hz simultaneously.

Figure 7 shows measurement of the sensor skin's response to normal stress at a single detection node. The measured resistance, i.e. the output of the sensing skin, increases monotonously with the normal contact pressure which can be inferred from the reading of the load cell, with a detection limit of 1.7mN.

Interestingly, the curve exhibits two distinct linear regions – The sensitivity above 1000KPa point is about 6 times bigger than below that. This deviation is presumably due to the viscoelastic behavior of the polymers – the strain is not only a function of the stress σ , but also a function of the time derivative of the stress $d\sigma/dt$.

Figure 8 shows the history of loading recorded in three separate tests. On the plot, all the curves exhibit two different slopes under and above the dashed line, indicating different time derivatives $d\sigma/dt$ – 796.89KPa/s and 439.98KPa/s for before and after the 1000KPa point, respectively. The difference in the rate of applied stress can be used to explain the distinct sensitivities in Figure 7 for lower and higher ranges.

While viscoelasticity is universal to polymeric materials, it is one property of the human skin too – the touch sensation is a function of both the magnitude and the rate of the stress. However, a verifiable model is yet to be developed for this complex mechanical – electrical coupling system to achieve a close form analytical solution for describing the complete stress response of the sensing skin.

Figure 8: The history of loading indicating two distinct zones where the rates of stress are different

CONCLUSIONS

We have demonstrated the concept, design, fabrication process and characterization results of a tactile sensing skin which is constructed by all-stretchable constituents. The sensing skin capable of capturing the contact events has distributed detection nodes, and is able to conform to the curved surface of a prosthetic limb part.

Major unique contributions of this work are summarized below:

- (1) Construction of a flexible touch sensing skin consisting of multiple stretchable constituents;
- (2) Initial feasibility studies for integrating such sensing skin on prosthetic limb parts;
- (3) Systematic characterization results of stretchable elements under loading conditions.

Future improvement includes complete integration with prosthetic limbs and comprehensive device characterization, contact imaging, analytical modeling and verification of the viscoelastic behavior of the sensing skin, etc.

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