MULTIFUNCTIONAL WEARABLE FINGERTIP ELECTRONICS

BY

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THESIS

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This thesis describes the device designs, fabrication methods and the results of simulations and tests for a class of electronics capable of integration onto the inner and outer surfaces of thin, elastomeric sheets in closed-tube geometries, specially formed for mounting on fingertips. Multifunctional systems of this type allow electrotactile stimulation with electrode arrays multiplexed using silicon nanomembrane diodes (Si NM), high-sensitivity strain monitoring with Si NM gauges, and tactile sensing with elastomeric capacitors. Experiments have been conducted for each of the devices to demonstrate the expected functionalities. Analytical calculations and finite element modeling of the mechanics quantitatively capture the key behaviors during fabrication/assembly, mounting and use. The results provide design guidelines that highlight the importance of the NM geometry in achieving the required mechanical properties. This type of technology could be used in applications ranging from human–machine interfaces to ‘instrumented’ surgical gloves and many others.
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1.1 Background: Electrotactile stimulators and tactile sensors

Electrotactile stimulation allows information to be presented through the skin, as an artificial sensation of touch, commonly perceived as a vibration or tingling feeling [1, 2]. Such responses manifest through the excitation of cutaneous mechanoreceptors as a result of passage of a suitably modulated local electrical current into the tissue [3]. Developed originally in the 1950’s and further advanced in the 1970’s, electrotactile stimulation has been traditionally explored for programmable braille readers and displays for the visually impaired as well as for balance control in individuals who suffer from vestibular disorders [4-8].

Different forms of electrotactile display have been studied, including single element display, which can present information to the skin by variations in intensity, frequency, or both; one-dimensional display, which consists a row of two or stimulation points that can present spatial information more naturally; and two-dimensional display, or called a tactile vision substitution (TVS), which is a two-dimensional matrix of stimulators that can display spatial information to the skin similarly to the way the eye presents spatial information to its retina [1]. The wearable system described in this

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1 This chapter includes previously published material from: M. Ying, A.P. Bonifas, N. Lu, Y. Su, R. Li, H. Cheng, A. Ameen, Y. Huang and J.A. Rogers, “Silicon Nanomembranes for Fingertip Electronics”, 2012 Nanotechnology 23
thesis incorporates a two-dimensional electrotactile stimulator matrix to accommodate applications such as programmable braille display and potentially vision substitution (with high-density electrode array).

Electrical current is passed to the skin by electrodes. And electrode parameters, together with the waveforms, determine the types and the quality of the sensation, as well as the level of skin irritation. Different electrode materials and geometries have been therefore studied [1]. The device described in this thesis uses gold coaxial electrode pairs, each of which consists of an active center electrode insulated from a larger annular surrounding dispersive electrode for the return current path.

Tactile sensors, on the other hand, measure pressure created by physical contact, in a way that provides complementary information for potential use in feedback loops with the electrotactile process.

Electrotactile stimulators and tactile sensors are of interest as bi-directional information links between a human operator and a virtual environment, in a way that could significantly expand function in touch-based interfaces to computer systems, with applications in simulated surgery, therapeutic devices, robotic manipulation and others [4, 9-12]. Additional classes of sensors that can be important in this context include those for motion and temperature.

Unlike many of the previously reported devices being bulky and built on rigid substrates, incorporating such technologies into a conformal, skin-like device capable of
intimate, non-invasive mounting on the fingertips might represent a useful achievement. Recent advances in flexible and stretchable electronics create opportunities to build this type of device [13-17].

1.2 Overview

This thesis reports materials, fabrication strategies and device designs for ultrathin, stretchable silicon-based electronics and sensors that can be mounted on the inner and outer surfaces of elastomeric closed-tube structures for integration directly on the fingertips. It also discusses the mechanical properties of the system and demonstrates the expected functions of each device.

Chapter 2 discusses the design for each of the devices, including the device structure, layout and geometries, and mechanic designs, which make the device capable of accommodating large strains induced not only by natural deformations of the tubes during use, but also during a critical ‘flip-over’ step in the fabrication process. This chapter then reports the fabrication process for each device and the finger tube. Chapter 3 introduces the mechanical models of the devices, discusses the maximum strain during the ‘flip-over’ process using both results from analytical calculations and finite element modeling (FEM), then explains the Neutral-Mechanical-Plane theory and how the design optimizations are achieved. Chapter 4 reports the experiment set-ups and the experiment results that demonstrate the expected functionality of each device.
individually. Chapter 5 concludes the results and discusses future challenges and potential study directions. Additionally, Appendix A reports and summarizes some other unpublished and ongoing projects that I have been working on.
Multifunctional fingertip devices that include electrotactile electrode arrays multiplexed with Si nanomembrane (NM) diodes, strain sensors based on Si NM gauges, and tactile sensor arrays that use capacitors with low-modulus, elastomeric dielectrics are first fabricated separately on silicon wafers, and then transfer-printed onto the inner and outer surfaces of a elastomeric closed-tube structure (the finger tube), which integrates the electronics and sensors directly on fingertips.

Large strains are induced not only by natural deformations of the tubes during use, but also during a critical step in the fabrication process in which the tubes, specially formed to match the shapes of fingertips, are flipped inside-out. This ‘flipping-over’ process allows devices initially mounted on the outer surface of the tube to be reversed to the inner surface, where they can press directly against the skin when mounted on the fingers. Therefore, the active components and interconnects need to incorporate advanced mechanics designs that are capable of accommodating these large strains and avoiding plastic deformation or fracture.

Ecoflex is chosen to be the substrate (finger tube) material. It is an attractive material for this purpose because it has a low modulus (~60 kPa) and large fracture

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2 This chapter includes previously published material from:
strain (~900%). The former allows soft, intimate contact with the skin; the latter enables the ‘flipping-over’ process referred to previously, and described in quantitative detail in Chapter 3.

In this chapter, the design layout of each device (sensor or actuator array) and detailed fabrication process are presented and discussed.

2.1 Device Design

2.1.1 Multiplexed electrotactile stimulator array

![Figure 1](image)

**Figure 1.** (a) Schematic illustration of a multiplexed electrotactile array with serpentine mesh interconnects, with magnified diagram (right top) and image (right bottom) of a PIN Si NM diode (after flipping-over). (b) Multiplexed electrotactile array mounted on the finger tube (finger tube inner surface exposed, before flipping-over)

As is shown in Figure 1, the electrotactile stimulator is a 2x3 electrode pair array, which is compatible with the braille pattern display application. The electrodes use 600 nm thick layers of Au in a concentric design, consisting of an inner disk (400 µm radius)
surrounded by an outer ring (1000 µm radius) with a 250 µm wide gap between the two. Interconnects consist of 100 µm wide traces of Au in serpentine geometries (radii of curvature ~800 µm); these traces connect the electrotactile electrodes to Si NM diodes (lateral dimensions of 225 µm x 100 µm and thicknesses of 300 nm). Each unit cell consists of one diode and one electrotactile electrode pair. The diodes enable multiplexed addressing. Two layers of Au interconnects (200 nm and 600 nm thick), isolated by a 1.25 µm PI layer and connected through etched PI vias, establish a compact wiring scheme with overlying interconnects. The 600 nm thick Au interconnect layer allowed robust electronic contact though the PI vias.

The short dimensions of the diodes are designed to lie parallel to the flipping-over direction, to minimize strains in Si during this process. Also, active components (Si NMs) and thin gold interconnects are encapsulated in between two PI layers of the same thickness, which places the functioning device close to the neutral mechanical plane (NMP), thereby minimizes the induced strains in these brittle materials during bending and flipping-over. More detailed analysis can be found in Chapter 3.

The electrotactile stimulators are mounted on the inner side of the finger tube, where they are in intimate contact with the skin on fingertip and therefore can pass stimulation current effectively. With a 2x3 array layout, the device can be used to display letters in braille pattern or to guide finger movements by turning on different stimulation electrodes and giving directional information. With the same design
principle and fabrication process, the stimulator array can be scaled up and contain more channels for further applications – High density array for image display, for example.

2.1.2 Strain gauge arrays

![Figures 2](image)

**Figure 2.** (a) Schematic illustration of a strain gauge array with serpentine mesh interconnects (b) an image of a fabricated device (c) an SEM image of a portion of the device, with the Si NM gauge located in the dashed box.

Figure 2 shows the mask design of the strain gauge array and the actual fabricated device. Each device consists of four Si NMs (strips with lateral dimensions of 1 mm x 50 µm and thicknesses of 300 nm) electrically connected by 200 nm thick, 60 µm wide Au traces patterned in serpentine shapes (radii of curvature ~400 µm).
**Figure 3.** A hand model wearing the finger tube with strain gauge array

After fabrication the device is then transfer printed onto the outer surface of the finger tube, and at the location where the strain gauges are aligned with the knuckle, see Figure 3. Strain change can be detected and measure when the finger bends or moves from side to side. Move details are included in Chapter 4.
2.1.3 Capacitive tactile sensor array

![Figure 4](image)

**Figure 4.** (a) Schematic illustration of the tactile sensor array and its relative location on the finger tube substrate with the electrotactile stimulator array. (b) Actual fabricated devices.

As is shown in Figure 4, the tactile sensors use 200 nm thick Au electrodes and interconnects in the geometry of the electrotactile array but with the concentric electrode pairs replaced by single, disc-shaped electrodes (radii ~1000 μm). The tactile sensor array is mounted on the outer surface of the finger tube, and when transfer printed, its disc-shaped electrodes are aligned with the concentric electrode pairs of the electrotactile array, so that each channel forms a parallel plate capacitor with the Ecoflex substrate being the dielectric in between. When pressure is applied against the fingertip onto these tactile sensors (for example when the finger is touching a surface or grabbing a tool), the Ecoflex substrate will be compressed and the distance between the two
electrode plates will change. Therefore the capacitance change can be detected and measured, indicating the force applied on the fingertip. More details are included in Chapter 4.

2.2 Fabrication Process

Figure 5. Schematic illustration of the process for transfer printing an interconnected device structure from a substrate on which it is fabricated to an elastomeric sheet. (a) Interconnected sensors and electronics formed on a silicon wafer in an open mesh geometry are lifted onto the surface of a PDMS slab (i.e. stamp); (b) the backside of the mesh and the supporting PDMS stamp are coated with a thin layer of SiO2 and then pressed onto an elastomeric sheet (Ecoflex); (c) removing the PDMS completes the transfer; (d) materials legend.

Figure 5 schematically illustrates steps for integrating devices in stretchable, interconnected geometries with elastomeric substrates, following adapted versions of procedures described elsewhere [13, 18]. In general, the fabrication uses a Si wafer with
a 100 nm thick coating of polymethylmethacrylate (PMMA) as a temporary substrate for the initial parts of the process. A layer of polyimide (PI; 1.25 mm thick) formed by spin coating a poly (amic acid) precursor and baking in an inert atmosphere at 250°C, serves as the support for the devices. Electronically active materials are deposited (e.g. metallization) or transfer printed (e.g. Si NMs) onto the PI and patterned by photolithography and etching. Another layer of PI (1.25 µm thick) spin cast and cured on top of the device layers provides encapsulation and locates the devices near the neutral mechanical plane (NMP). Next, patterned reactive ion etching through the entire multilayer stack (i.e. PI/devices/PI) defines an open mesh structure. This same process removes PI in regions of the electrotactile stimulation electrodes, to allow direct contact with the skin. Immersion in an acetone bath washes away the underlying PMMA, thereby allowing the entire mesh to be lifted off, in a single piece, onto the surface of a flat slab of polydimethylsiloxane (PDMS), using procedures described previously [19, 20]. Evaporating a layer of SiO2 onto the mesh/PDMS and exposing the silicone target substrate (Ecoflex 0030, Smooth-On, Inc.) to UV-ozone (to create reactive –OH groups the surface) enables bonding between the two upon physical contact [21]. (Low pressures avoid contact between the PDMS and the finger-tube, thereby allowing bonding only to the mesh). The SiO2 adhesion layer does not serve any electronic function. Removal of the stamp completes the transfer process, as shown in Figure 5c.
2.2.1 Silicon Transfer-printing

Figure 6. Schematic of silicon transfer printing. (a) silicon on insulator (SOI) substrate; (b) RIE etch release holes (3 µm) in Si layer; (c) wet etch (buffered oxide etch) of SiO2 layer to release Si layer; (d) PDMS stamp pressed into contact with Si; (e) Si transfer to PDMS stamp upon removal; (f) PDMS stamp with transferred Si pressed onto PI layer; (g) After heating at 150°C for 4 min, Si transferred to device upon stamp removal.

Both the multiplexed electrotactile stimulator array and the strain gauges use Si NMs as their active function components – Multiplexing diodes as in the electrotactile stimulator array and the measuring component as in the strain gauges. In order to incorporate Si NMs in these devices, a silicon film is released from the silicon on insulator (SOI) substrate, then transfer printed onto the target wafer, and finally patterned. Steps are as following and shown in Figure 6:

(1) Create holes (3µm dia., spacing 30µm) for releasing Si film:
i. Spin coat PR Shipley S1805 (3000rpm, 30s), pre-bake (110°C, 1min),
align mask and expose, develop with MIF327 (9s), post-bake (110°C,
3min).

ii. Etch Si with RIE (50 mtorr, 40 sccm SF6, 100 W, 1min).

(2) Undercut oxide layer of SOI:

i. Immerse wafers in HF solution for 15~20min until the Si layer is
detached from the substrate.

(3) Pick up the Si film from the SOI wafer with a PDMS stamp.

(4) Prepare target Si wafer:

i. Spin coat Si wafer with polymethylmethacrylate (PMMA, 3000rpm,
30s, ~100nm), cure at 180°C for 1.5min.

ii. Spin coat polyimide precursor (4000rpm, 30s) and partially cure at
150°C for 40sec.

(5) Transfer Si to target Si wafer:

i. Press the stamp into contact with the target wafer and apply force
with hands for 10s.

ii. Put stamp and target wafer on a hotplate at 110°C and slowly release
the stamp when thermal expansion of the stamp is observed.

iii. Put target wafer (now with Si film) on hotplate at 150°C for another
5min and remove PR with acetone (2 s).
iv. Bake in an inert atmosphere at 250°C for 1hr.

(6) Si isolation

i. Pattern PR AZ5214.

ii. Etch exposed Si with RIE (50 mtorr, 40 sccm SF6, 100 W, 1min) and strip PR with acetone.

2.2.2 Multiplexed electrotactile stimulator array

Figure 7. Schematic of the fabrication process for electrotactile stimulators. (a) silicon substrate; (b) spin coat 100 nm sacrificial PMMA; (c) spin coat/250°C bake 1.2 µm polyimide; (d) transfer of Si layer with PIN diodes (release holes not shown); (e) RIE isolation of Si nanomembrane PIN diodes and Au evaporation/patterning; (f) spin coat/250°C bake 1.2 µm polyimide; (g) contact vias for diodes formed in PI with O2 RIE; (h) Au evaporation/patterning; (i) spin coat/250°C bake 1.2 µm polyimide; (j) O2 RIE to form polyimide mesh structure and expose electrotactile electrodes.
Figure 7 shows the schematic of the fabrication process for electrotactile stimulators. Detailed steps are as following:

(1) Cut 1’x1’ SOI wafers ((110), 300nm Si) and clean with acetone and IPA.

(2) Form a 900nm layer of SiO2 by PECVD as p-dope diffusion mask.

(3) Pattern diffusion mask:
   i. Pattern photoresist (PR) AZ5214: Spin coat PR AZ5214 (3000rpm, 30s), pre-bake (110°C, 1min), align mask and expose, develop with MIF327 (40s), post-bake (110°C, 3min).
   ii. Wet etch with buffered oxide etchant (BOE) (NH4F: HF=6:1) for 1.5 min and remove PR with acetone.

(4) P-type doping:
   i. Clean wafers with Nano-strip™ (Cyantek), place next to boron doping source, and put into furnace (1000°C) for 30min.
   ii. Etch SiO2 mask completely with HF (30sec), and form another 900nm layer of SiO2 by PECVD as n-dope diffusion mask.
   iii. Pattern diffusion mask: Same as step (3).

(5) N-type doping:
   iv. Clean wafers with Nano-strip™, place next to phosphorous doping source at 1000°C for 10min.
   v. Etch SiO2 mask completely with HF (30sec).
(6) Silicon transfer printing and diode isolation: **See steps in 2.2.1**

(7) 1st Au interconnect layer:
   
i. Deposit Cr (5 nm)/Au (200 nm) with electron beam evaporator.
   
ii. Pattern PR AZ5214.
   
iii. Wet etch Au and Cr.
   
iv. Strip PR with acetone.

(8) PI insulation layer with vias:
   
i. Spin coat polyimide precursor (4000rpm, 30s).
   
ii. Prebake on hotplate (150°C, 5min).
   
iii. Bake in an inert atmosphere at 250°C for 1hr.
   
iv. Spin coat PI with PR AZ4620 (3000rpm, 30s), pre-bake (110°C, 1min),
   
   align via mask and expose, develop with 3:1 diluted MIF400 (40s).
   
   v. Etch exposed polyimide with RIE (100W, 150mTorr, 20sccm O₂, 20min).
   
   vi. Strip PR with acetone.

(9) 2nd Au interconnect layer:
   
i. Deposit Cr (10 nm)/Au (600 nm) with electron beam evaporator.
   
ii. Pattern PR AZ5214.
   
iii. Wet etch Au and Cr.
   
iv. Strip PR with acetone.
(10) Final PI encapsulation and etch:

i. Form PI layer: Same as steps (8).i – (8).iii.

ii. Pattern PR AZ4620.

iii. Etch exposed polyimide with RIE (100W, 150mTorr, 20sccm O2, 50min) to form PI mesh structure.

iv. Strip PR with acetone.

(11) Transfer printing

The electrotactile stimulators are transfer printed onto the inner surface of the finger tube eventually, where they can press directly against the skin when mounted on fingers. For steps see 2.2.5
2.2.3 Strain gauge array

Figure 8 shows the schematic of the fabrication process for silicon strain gauges.

Detailed steps are as following:

1. Cut 1’x1’ (110) SOI wafers (300 nm Si) and clean with acetone and IPA.

2. P-type doping: same as step 2.2.2(4) with a 4 min doping time.

3. Silicon transfer printing and strain gauge isolation: See steps in 2.2.1

4. Au interconnect layer: same as step 2.2.2(7)

5. Final encapsulation: same as step 2.2.2(10) with 30 min O2 RIE.

6. Transfer printing:
The Strain gauge arrays are eventually transfer printed onto the outer surface of the finger tube at the knuckle position, where they can sense the strain induced by finger motions. For steps see 2.2.5

2.2.4 Contact sensor array

Figure 9. Schematic of the fabrication process for tactile electrodes. (a) silicon substrate; (b) spin coat 100 nm sacrificial PMMA; (c) spin coat/250°C bake 1.2 μm polyimide; (d) Au evaporation/patterning; (e) spin coat/250°C bake 1.2 μm polyimide; (f) O2 RIE to form polyimide mesh structure.

Figure 9 shows the schematic of the fabrication process for the contact sensors (outer side half, inner side half is the electrotactile stimulator array). Detailed steps are as following:

1. Cut 1’x1’ Si wafers and clean with acetone and IPA.
2. Spin coat PMMA (3000 rpm, 30s) as sacrificial layer.
3. Form polyimide layer as substrate: Same as 2.2.2(8).
4. Au interconnect layer: same as step 2.2.2(7).
5. Final encapsulation: same as step 2.2.3(5).
(6) Transfer printing to overlay with electrotactile electrodes: same as step 2.2.3(6).

2.2.5 Finger tube

The Ecoflex substrates, which are referred to as finger-tubes, adopt three-dimensional forms specifically matched to those of fingers on a plastic model of the hand. The fabrication involves pouring a polymer precursor to Ecoflex onto a finger of the model and curing at room temperature for 1 hour, to create a conformal sheet with ~125 μm thickness. Pouring a second coating of precursor onto this sheet and curing for an additional 1 hour doubles the thickness; repeating this process 4 times results in a thickness of ~500 μm. Removing the Ecoflex from the model and completing the cure by heating at 70°C for 2 hours forms a free standing structure, i.e. a finger-tube, like the one illustrated in Figure 10(a).
2.2.6 Transfer-printing and flip-over

**Figure 10.** Process for transfer printing a multiplexed array of electrotactile stimulators in a stretchable, mesh geometry onto the inner surface of an elastomeric finger-tube. (a) casting and curing an elastomer precursor on the finger of a model hand yields a thin (~500 mm thick), closed-form membrane, i.e. a finger-tube; (b) a PDMS stamp (here, backed by a glass microscope slide) delivers the electrotactile device to the outer surface of this finger-tube, while compressed into a flattened geometry; (c) electrotactile array on the outside of the freestanding finger-tube; (d) turning the tube inside out relocates the array on the inner surface of the finger-tube, shown here at the midway point of this flipping process.

After being fabricated on silicon substrates, the device mesh structures are delivered to the outer and inner surfaces of the finger-tube by transfer printing, while pressed into a flattened geometry (Figure 10b). Steps are as following:

1. Immerse device in heated acetone bath (100°C) to undercut PMMA.

2. Press PDMS stamp into contact with the device and quickly remove to transfer device onto the stamp.
(3) Deposit Cr (5 nm)/SiO2 (20 nm) with e-beam evaporator.

(4) Ultra-violet/ozone(UV-O) treat the target substrate (Ecoflex finger tube) for 4min.

(5) Press the PDMS stamp onto Ecoflex and remove stamp slowly.

When transfer printing the electrotactile stimulator array, the finger tube is first flipped inside-out to expose the inner surface. After transfer printing, the entire integrated system is then flipped over again, to move the device from the outer to the inner surface of the tube, as shown in Figure 10c,d.

Eventually, the fingertip multifunctional devices incorporate electrotactile stimulators on the inside, and strain gauge arrays and tactile sensors on the outside.
CHAPTER 3: MECHANICAL MODELING

This chapter is based on the collaboration work with Department of Mechanical Engineering and Department of Civil and Environmental Engineering, Northwestern University.

3.1 Mechanics modeling of the flip-over process

The device designs described previously have the advantage that they are conformal to the finger, in a way that naturally presses the electronics on the interior surface of the finger tube (in this case the electrotactile stimulating electrodes) into intimate contact with the skin. The flipping-over process represents a critical step, enabled by careful design of the mechanics in the device mesh. Quantitative mechanics modeling, including analytical calculations and finite element modeling (FEM) provides important insights into these design layouts that avoid plastic deformation or fracture.

3.1.1 Maximum strain analysis

The finger-tube can be approximated as a self-equilibrated, axisymmetric tube with two-dimensional symmetry. Figure x schematically shows a finger tube with a thickness of $t_{sub}$, bent back on itself on a finger model with a radius of $R_{radial}$, at a

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<sup>3</sup> This chapter includes previously published material from:
midway point during the flipping-over process. AB represents the cylindrical portion in contact with the surface of plastic hand; the outer surface DE is also cylindrical; transition between the two can be approximated by a semi-circle BC (with radius R1 to be determined) and a sinusoidal curve CD (with half wavelength L to be determined). The linear elastic shell theory gives the bending energy and the membrane energy. Minimization of the total energy then gives R1 and L, which determine the resulting shapes.

\[ y = R_1 + \frac{1}{2} t_{sub} + \left( R_1 - \frac{1}{2} t_{sub} \right) \cos \left( \frac{\pi x}{L} \right) \]

\[ \sinusoidal \ curve \ CD \]

\[ \text{Schematic of the finger tube model at a midway point of the flipping-over process, showing the geometries determined by linear elastic shell theory} \]

\[ \text{Figure 11} \]

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4 This figure was created by Y. Su, R. Li and H. Cheng, Department of Mechanical Engineering and Department of Civil and Environmental Engineering, Northwestern University.
Figure 12. (a) Calculated (analytical and FEM) profiles of an Ecoflex finger-tube during bending associated with flipping the tube inside out, showing the relationship between the radius ($R_{radial} = 7.5$ mm) of the tube and the minimum bending radius ($R_{axial}$); (b) FEM results for maximum strains on the inner and outer surfaces during this process;

Figure 12(a) shows analytical and FEM results of the finger tube shapes for tube radius $R_{radial} = 7.5$ mm and thickness $= 500$ µm. The minimum axial radius of curvature ($R_{axial}$) of 596 µm, as indicated in figure 12(a), defines the location of maximum induced strain as the tube is flipped over. As is shown in the color map of figure 12(b), in this configuration, the maximum tensile strain, which is on the inner surface, is $\varepsilon_{tensile} = 34.4\%$; and the maximum compressive strain, which is on the outer surface, is $\varepsilon_{compressive} = -49.5\%$. These calculated results also agree well with the FEM results ($\varepsilon_{tensile} = 35.1\%$ and $\varepsilon_{compressive} = -46.9\%$). The device mesh structures must, therefore, be able to accommodate strains in this range. This requirement is non-trivial for systems like the ones described here, due to their incorporation of brittle materials such as

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5 This figure was created by Y. Su, R. Li and H. Cheng, Department of Mechanical Engineering and Department of Civil and Environmental Engineering, Northwestern University.
silicon (fracture strain ~ 1%).

The multiplexed electrotactile arrays are modeled as a composite beam with multiple layers. The bending moment and membrane force obtained from the above analytical model are imposed on the multiplexed electrotactile arrays. This gives the analytical expressions of the maximum strain in Si and Au, which are validated by FEM. The maximum calculated strains turned out to be only 0.051%, 0.10%, and 0.040% for the Au, PI, and the Si, respectively, as shown in figure 13.

![Figure 13](image)

**Figure 13.** Schematic cross sectional illustrations of two regions of the device, with the position of the NMP indicated by a dashed red line, and analytical results for the maximum strains during the flipping-over process

3.1.2 Design Optimizations

The optimizations made in the design of circuit layouts lead to the small amount of strains in the device (only 0.051%, 0.10%, and 0.040% for the Au, PI, and the Si, respectively), while the strain in the Ecoflex substrate is fairly large (30%~40%). As an
example, the multiplexed electrotactile array is designed in a mesh configuration with narrow serpentine interconnects, and the short dimensions of the diodes lie parallel to the flipping-over direction, which minimizes strains in the Si during this process.

**Figure 14.** Maximum strain in the Si NM diode and $h_{NMP}$ (the offset between the neutral mechanical plane and the lower surface of the Si NM) as a function of thickness of the Si NM.

The computed position of the neutral mechanical plane (NMP) also appears in Figure 13. Since the moduli of the device layers are several orders of magnitude larger than that of Ecoflex, the location of the NMP plane is largely independent of the Ecoflex. Appropriate selection of the thicknesses of the PI layers allows the NMP to be positioned at the location of the Si NMs, thereby minimizing the induced strains in this brittle material [21, 22]. The thicknesses of the Si NM diodes influence the maximum strains that they experience, as shown in analytical calculations of Figure 14. A
minimum occurs at the thickness that places the NMP at the shortest distance from the Si NM diode (i.e. \( h_{\text{NMP}} \)). The position of this minimum can also be adjusted by changing the thicknesses of the PI layers, for example. Further reductions in strain can be realized by reducing the lengths of the devices. Implementing designs that incorporate these considerations and together with use of interconnects with optimized serpentine layouts ensures robust device behavior throughout the fabrication sequence. For example, Figure 15 shows negligible change in the I-V characteristics (Agilent 4155C semiconductor parameter analyzer) of a Si NM diode before and after the flipping-over process.

![Figure 15](image)

**Figure 15.** I-V characteristics of a Si NM diode before and after flipping-over.

3.2 Mechanical modeling of the strain gauges

Figure 16 shows a set of straight, uniformly doped Si NMs as strain gauges
addressed with interconnects in a mesh geometry. The FEM calculations summarized in Figure 16 reveal the strain profiles in a 1X4 array of gauges (vertical strips; the yellow dashed box in the upper inset highlights an individual device) on Ecoflex, under a uniaxial in-plane strain of 10%. These results show that the overall strain is mostly accommodated by changes in the shapes of the serpentine interconnects and, of course, the Ecoflex itself. The Si NM gauges experience strains ($\sim 10^{-3}$) that are ten times lower than the applied strain, as shown in the inset in figure 16.

![Figure 16](image)

**Figure 16**. FEM results of the maximum principle strain for a 1 x 4 array of gauges (straight, vertical structures near the top of the serpentine interconnect mesh) due to an overall 10% strain applied along the longitudinal (y) direction. The upper inset shows the strains in the gauge highlighted by the yellow dashed box. The lower inset provides an image of a fabricated device with a layout that matches that of the FEM results.

The ability to use Si NMs as high performance strain gauges in stretchable forms results from the strong piezoresistance properties of Si, combined with serpentine

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6 This figure was created by Y. Su, R. Li and H. Cheng, Department of Mechanical Engineering and Department of Civil and Environmental Engineering, Northwestern University.
layouts. These characteristics, taken together, determine the fractional change in resistance per applied strain. The associated effective gauge factor ($GF_{eff}$) can be related to the intrinsic gauge factor of a silicon gauge:

$$GF_{si} = \frac{\Delta R}{\Delta R \cdot \varepsilon_{si}}$$  \hspace{1cm} (3.1)

where $\Delta R$ is the change in resistance, $R$ is the initial resistance, and $\varepsilon_{si}$ is the strain in the silicon, by the following expression:

$$GF_{eff} = GF_{si} \left( \frac{\varepsilon_{si}}{\varepsilon_{app}} \right)$$ \hspace{1cm} (3.2)

where $\varepsilon_{app}$ is the strain applied to the overall, integrated system. The designs reported here yield values of $\varepsilon_{si}/\varepsilon_{app}$ that are much smaller than one, specifically to avoid fracture-inducing strains in the Si during fabrication, mounting and use over physiologically relevant ranges of strain.

### 3.3 Mechanical modeling of the tactile sensors

The concentric electrodes of the electrotactile stimulators on the inner surface and the disc-shape electrodes of the tactile sensors on the outer surface form pairs of parallel capacitors. The capacitance change is related to the applied pressure that results in the decrease of the thickness of Ecoflex dielectric

$$\frac{\Delta C}{C_0} = \frac{P}{E_{Ecoflex} - P},$$ \hspace{1cm} (3.3)
where $E_{\text{Ecoflex}} = \frac{(1-\nu)E}{(1+(1-2\nu))}$ is the effective modulus of Ecoflex dielectric under uniaxial stretching, and $E=60$ kPa is the Young’s modulus of Ecoflex. It will be shown in chapter 4 that Eq. (3.3) agrees well with experiments.

For an applied tensile strain $\varepsilon_{\text{applied}}$, the strain in the Ecoflex dielectric between electrodes is related to the tensile stiffness $(EA)_{\text{system}}$ of the system and tensile stiffness $(EA)_{\text{electrodes}}$ of the electrodes by $\varepsilon_{\text{applied}} (EA)_{\text{system}} / (EA)_{\text{electrodes}}$. The capacitance change of a single element of the pressure sensor array is also determined by the decrease of the thickness of the Ecoflex dielectric, and is given by

$$\frac{\Delta C}{C_0} = \frac{(EA)_{\text{system}} - (EA)_{\text{electrodes}} \nu_{\text{applied}}}{(EA)_{\text{electrodes}}}.$$  (3.4)
CHAPTER 4: FUNCTIONALITY TESTS

Each of the three devices on finger tube has been tested individually to demonstrate the expected functionalities.

4.1 Multiplexed electrotactile stimulator array

4.1.1 Expected functionality

(1) Electrode pairs are in intimate contact with fingertip skin and able to pass suitably modulated electrical current to excite cutaneous mechanoreceptors, so that the subject gets perception of touch;

(2) Multiplexed addressing with the diodes;

(3) Diodes are stable in the range of applied voltage and current that is sufficient for electrotactile stimulation;

(4) Electrical isolation of adjacent channels.

4.1.2 Experiment setup

The electrotactile stimulator array has been tested on both dry and hydrated human thumbs. A saline-based conductive gel was used to hydrate the thumb before the finger tube was put on. The device was connected to the external electronics via a flexible anisotropic conductive film (ACF). In the dry-thumb test, the stimulation current was applied using a constant voltage source, and modulated by a Python program and a relay-based (a 4x6 relay array generating square wave pulse series and
controlling each channel in the stimulator array) control circuit. Figure 17 shows the control circuit. In the hydrate-thumb test, a probe station was used to apply the sweeping voltage to the stimulation channel and to read out the passing current.

**Figure 17.** Control circuit for the dry-thumb test. Red dash line square shows the 4x6 relay array

4.1.3 Results and discussion

Experimental results demonstrate expected functionality in the electrotactile stimulator arrays. Figure 18(a) shows the perception of touch on a dry human thumb as a function of voltage and frequency, applied between the inner dot and outer ring electrodes (Figure 1a). Stimulation used a monophasic, square-wave with 20% duty cycle, generated the setup described previously. The inset provides an image of a device, with connection to external drive electronics via a flexible anisotropic conductive film (ACF). The required voltage for sensation decreases with increasing frequency,
consistent with equivalent circuit models of skin impedance that involve resistors and capacitors connected in parallel. The absolute magnitudes of these voltages depend strongly on the skin hydration level, electrode design, and stimulation waveform [23]. Figure 18(b) shows I-V characteristics of an electrotactile electrode pair while in contact with a hydrated human thumb, measured through a multiplexing diode. At high positive voltages, the resistance of the diode is negligible compared to the skin; here, the slope of the I-V characteristics yield an estimate of the resistance of the skin-electrode contact plus the skin. The value (~40 kΩ) is in a range consistent with measurements using conventional devices [24, 25]. The diode is stable to at least 20 V, corresponding to currents of 0.25 mA, which is sufficient for electrotactile stimulation on the skin and tongue [1, 2, 10].

**Figure 18.** Mechanics and electrical characteristics of a 2 x 3, multiplexed electrotactile array on a finger-tube. (a) Voltage required for electrotactile sensation as a function of stimulation frequency. Inset: electrotactile array on human finger during experiments; (b) I-V characteristics of multiplexed electrotactile electrodes in contact with a human thumb.
Figure 19. (a) circuit diagram of the diode multiplexing scheme; (b) function table showing inputs for addressing each of the six channels (H = High; L = Low).

These diodes enable multiplexed addressing, according to an approach that appears schematically in Figure 19(a). Each unit cell consists of one diode and one electrotactile electrode pair. Figure 19(b) presents a table of the inputs required to address each of the six electrotactile channels. For example, channel SDA can be activating by applying a high potential (+5 V) to inputs A and E and a low potential (0 V) to inputs B, C, and D, thereby yielding a +5 V bias across the outer ring (+5 V) and inner ring electrode (0 V) of this channel. This configuration forward biases the Si NM diode, which results in stimulation current, as shown in Figure 18(b). At the same time, channels SEB and SEC experience a bias of -5 V across the electrodes but in these cases the Si NM diodes are reverse biased, thus preventing stimulating current. Channels SDB, SDC, and SEA have the same potential on the inner and outer electrodes, resulting
in zero bias. Electrical isolation of adjacent channels is a consequence of inner to outer electrode separations (250 µm) that are small compared to the distances between channels (6000 µm). Advanced multiplexing schemes that use several diodes per stimulation channel, or active transistors, are compatible with the fabrication process and design principles outlined here.

4.2 Strain gauge array

4.2.1 Expected functionality

(1) Applied strains mostly accommodated by the elastomeric substrate and the serpentine interconnects, so as to avoid fracture of the Si NMs;

(2) Large enough gauge factor to identify strain change induced by finger motions;

(3) Ability to decouple and identify strains induced by different finger motions (e.g. bending, moving from side to side, etc.);

4.2.2 Experiment setup

The strain gauge device is connected to the external electronics via an ACF cable. In the calibration test (from which we got the “strain-resistance change” curve and the effective gauge factor), a stretcher shown in figure 20 was used to apply and to measure the overall strain. And the resistance change ΔR was evaluated at 1V, using an Agilent 4155C semiconductor parameter analyzer.
4.2.3 Results and discussions

Figure 20. Stretcher used in the calibration test of strain gauge arrays to apply and to measure overall strain

Figure 21. Experimentally measured and analytically calculated changes in resistance for a representative Si NM strain gauge as a function of applied strain along the longitudinal direction. The inset provides an SEM image of a portion of the device, with the Si NM gauge located in the dashed box

Figure 21 shows experimentally measured values of $\Delta R$ as a function of $\varepsilon_{\text{app}}$, which corresponds to $GF_{\text{eff}} \sim 1$. By fitting the experimental and FEM results to Figure x,
the GF_{Si} is ~ 95, consistent with a recent report on Si NM strain gauges, with otherwise similar designs, on flexible sheets of plastic [26]. We emphasize that device design parameters, such as the size of the gauge and the dimensions of the serpentine interconnects, enable engineering control over GF_{eff} from values as large as GF_{Si} to those that are much smaller, with a correspondingly increased range of strains over which measurements are possible.

**Figure 22.** (a) images of a strain gauge array on a finger-tube mounted on the thumb, in straight (I) and bent (II) positions; (b) change in resistance of a representative gauge during three bending cycles (black) and side-to-side motion (red); (c) images of a strain gauge array on a thin, elastomeric sheet laminated onto the metacarpal region of the thumb in straight (III) and sideways deflected (IV) positions; (d) change in resistance of gauges at two ends of the array during three cycles of side-to-side motion.
Figure 22(a) shows a strain gauge array on a finger-tube located near the knuckle region of the thumb, in straight (I) and bent (II) positions. Upon bending, the gauges experience tensile strain, resulting in an increase in resistance, as shown for three bending cycles in Figure 22(b). The relative resistance changes suggest that the strain associated with bending reaches ~6%. As expected, side-to-side motions induced no changes (As the red curve in Figure 22(b) shows). Figure 22(c) shows a similar array on a thin sheet of Ecoflex, mounted near the metacarpal region of the thumb. Here, the device adheres to the skin by van der Waals interactions, similar to mechanisms observed in epidermal electronic systems [13]. The images in Figure 22(c) correspond to the thumb in straight (III) and sideways deflected (VI) positions. The changes in resistance for the two gauges on opposite ends of the 1 x 4 array for three side-to-side cycles of motion appear in Figure 22(d). For each cycle, the change in resistance of the rightmost gauge indicates compressive strain; the leftmost indicates corresponding tensile strain. The results suggest that arrays of gauges can be used to identify not only the magnitude but also the type of motion.

4.3 Contact sensor array

4.3.1 Expected functionality

(1) When pressure is applied, the thickness of the elastomeric dielectric changes effectively that the capacitance changes proportionally to the applied pressure;
Sensors insensitive to the thickness change induced by the in-plane strain (stretching of the substrate).

4.3.2 Experiment setup

The tactile sensor device is connected to the external electronics via an ACF cable. Pressure is applied with a series of weights (ranging from 50g to 500g), mounted on a platform with constant contact area. And the capacitance is measured with an Agilent E4980A LCR meter.

4.3.3 Results and discussions

This device exploits changes in capacitance associated with opposing electrodes on the inner and outer surfaces of the Ecoflex. Applied pressure decreases the thickness of the Ecoflex, thereby increasing the capacitance of this structure. Here, layouts like those for the electrotactile devices serve as inner electrodes; a mirror image of this array mounted in an aligned configuration on the outer surface defines a collection of parallel plate capacitors with the Ecoflex as the dielectric. An array of such devices on the anterior surface of a model of the hand appears in Figure 23(a). Figures 23(b) and 23(c) show images of the inner and outer electrode arrays.
Figure 23. Tactile sensing with integrated capacitance sensors. (a) sensors on the anterior of the thumb; (b) inner electrodes for a 2 x 3 array of sensors (electrotactile electrodes); (c) outer electrodes for the same array; (d) measured and analytically calculated change in capacitance of a single sensor with applied pressure and tensile strain.

The relative change in capacitance with applied pressure for a representative device appears in Figure 23(d) (black symbols). Here, capacitance was measured as a function of pressure applied with a series of weights mounted on a platform with a constant contact area, taking care to minimize effects of parasitic capacitances and to
eliminate ground loops. Approximately linear behavior is observed over the range studied, consistent with simple mechanical models:

\[
\frac{\Delta C}{C_0} = \frac{P}{E_{\text{ecoflex}} - P},
\]

(4.1)

where \(\Delta C\) is the capacitance change, \(C_0\) is the initial capacitance, \(P\) is the applied pressure, and \(E_{\text{ecoflex}}\) is the effective Ecoflex modulus. This simple model assumes no electrostriction or strain induced changes in dielectric constant (Figure 23(d), black line). Due to the Poisson effect, the device also responds to in-plane strains (\(\varepsilon_{\text{applied}}\)), as shown in Figure 23(d) (red), consistent with the simple model:

\[
\frac{\Delta C}{C_0} = \left[\frac{(EA)_{\text{system}}}{(EA)_{\text{electrodes}}}\right] \nu \varepsilon_{\text{applied}},
\]

(4.2)

where the Poisson’s ratio \(\nu\) is 0.496, and \((EA)_{\text{system}}\) and \((EA)_{\text{electrodes}}\) are the tensile stiffnesses of the system and electrodes respectively. This type of technology provides a simple alternative to recently reported devices that offer similar functionality, but on flexible substrates, and based on conductive elastomers, elastomeric dielectrics, or compressible gate dielectrics in organic transistors. [14, 16, 18, 27, 28].
CHAPTER 5: CONCLUSIONS AND FUTURE DIRECTIONS

5.1 Conclusions

By integrating both stimulator (electrotactile stimulation array) and sensors (strain gauges and tactile sensors) onto the inner and outer surfaces of a skin-like substrate in closed-tube geometry, we are able to achieve a bi-directional human-machine interface that is easy to wear, capable of intimate, non-invasive mounting on the fingertips, and mechanically invisible.

Moreover, the results presented in this thesis establish some procedures and design rules for electronics and sensors that can be mounted conformally onto the fingers. Other appendages of the body can be addressed in similar manner. Furthermore, most of the considerations in mechanics and fabrication are agnostic to the specific device functionality or mounting locations. As a result, many of these concepts can be applied generally, to other types of systems and modes of use.

5.2 Future challenges and directions

The future challenges and study directions include:

(1) Wireless power supply and data transfer
Currently the device relies on ACF connection to external electronics for power supply, multiplexing control, and measurement readings, which to some extent limit the mobility of the operator when using the device. The studies and integration of wireless modules onto the finger-tube platform would add much to the convenience and flexibility of using the device and even make remote control and data transfer possible.

(2) Modeling and Optimization of stimulation electrodes and waveforms

For the electrotactile stimulation, current waveforms and electrode parameters determine the mode of information transferred to the user, the qualitative sensation, and the level of skin irritation [1]. Therefore the current distribution under stimulation electrodes of various sizes and geometries needs to be studied; the electrode-skin interface needs to be accurately modeled (e.g. hydrated and non-hydrated, etc.); and it might also be interesting and necessary to study how to differentially excite different afferent fiber types to achieve the desired information transfer [29]. Therefore a model is needed to predict which fibers are preferentially excited by different waveforms and electrode parameters. All of these studies then could lead to an optimized device design and waveform used for stimulation.

(3) Integration of additional classes of sensors/actuators

The elastomeric finger tube provides an open platform for multifunctional sensors and actuators to be integrated onto. Aside from the devices described in this
thesis, additional classes of sensors that can be important to look into include temperature sensors, hydration level sensors, and optical sensors that can measure blood oxygen level, etc. In the case of temperature and hydration level sensors, the substrate might also need to be engineered to be vapor permeable.
REFERENCES


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APPENDIX A: UNPUBLISHED AND ONGOING PROJECTS

In this appendix, I will summarize some other unpublished or ongoing projects/experiments that I have done during my thesis research.

A.1 Other designs and functionality tests for the electrotactile stimulator array

1. Other designs

The electrotactile stimulator array has been designed with other electrode shapes and geometries, for the purpose of testing and exploring the functionality of the device and optimizing the final design.

![Figure 24](image)

**Figure 24.** Two other designs for the electrotactile stimulator. (a) 11-channel concentric-ring electrodes design. (b) 4x6 high-density stimulator array design.

Figure 24(a) shows a stimulator design with one circular shape electrode in the center serving as the cathode, and other 10 other concentric rings surrounding the
cathode serving as anodes. The diameter of the center cathode is 0.5mm, and the diameter of ring-shape anodes varies from 2mm to 12mm. This design is to study the “size effect” of the stimulation electrodes and to optimize the anode-cathode distance (or the anode-cathode size ratio) – the stimulating current, and thus the stimulation is confined in the circular area by the outer ring anode. If this stimulation area is too small, the stimulation would be difficult to be perceived, however the area being larger would compromise the resolution of the stimulator array. This design is also a one-dimensional display that can present spatial information – The anode channels can be turned on one by one either from inside or from outside, and the propagation of stimulation signals could be perceived.

Figure 24(b) shows a high-density version (4x6) of the electrotactile stimulator array, and the insets show a single stimulation pixel with one pair of electrodes. The size of each pixel (the outer ring electrode) is 2mm in diameter. With the design the stimulation can be more accurate and more complicated patterns can be presented to the skin.

2. Two-point/Three-point discrimination test

This test was carried out to test the resolution of the electrotactile stimulator array, and to see if the operators can distinguish stimulations from multiple points.
A 3x2 stimulator array transfer-printed on the inner side of an Ecoflex finger tube was used. The outer ring electrode size of each stimulation pixel is 3mm in diameter and the spacing between pixels is 6mm, as shown in Figure 25. A 40 volts, 80Hz square wave voltage was used to stimulate. Four subjects participated in this test, and each of them was given a single-point stimulation at first then proceeded to two-point and three-point stimulation. All subjects could perceive the single-point stimulation and distinguish the two-point stimulation of any spacing (minimum 6mm). As for the three-point stimulation, only some certain patterns (stimulating channels relatively apart from each other) can be distinguished. See Figure 25.

3. EKG measurement with finger/toe tubes

In this experiment, the electrotactile stimulator arrays were used as single-channel Electrocardiography (EKG) sensors and connected to a bio-amplifier to monitor the electrical activity of the user’s heart.
Three elastomeric (Ecoflex) tubes with such device on inner side were put on the subject’s both thumbs (measuring EKG lead I) and one of the toes on right foot (ground), as shown in Figure 26(a). Figure 26(b) shows the result: EKG signal obtained with this method shows clear QRS complex and T wave, from which we can extract important information about the user’s heart activity (heart rate, etc.)

**Figure 26.** Measurement of EKG with finger tube device (a) Experiment set-up (b) EKG signal
This method provides a more convenient way of measuring EKG, and it also gives us inspiration on the potential applications of toe-tip electronics, which are non-visible, and less intrusive for normal life styles compared to fingertip electronics. They can be used not only as the third electrode for EKG sensing, but also for pacing (dancing sports training) and giving directions. Similar design with other kinds of sensors such as pressure sensors can also be used to monitor other activities (running speed for example).

A.2 Photo patternable PDMS substrate

In order to create patterns on polymer substrates to reduce effective modulus and to increase vapor permeability, we tried to mix photo-initiator Benzophenone with PDMS precursor, so that free radicals are generated in the UV exposed part of PDMS during the photolithography process, which prevent the polymer from curing [30]. Detailed steps are as following:

1. Mix PDMS precursor (elastomer + curing agent) with wt. 5% Benzophenone;
2. Spin coat the mixture onto glass substrate at 500rpm for 1min (~200µm in thickness);
3. UV expose: 15min with UV intensity 27mW/cm²;
4. Bake immediately for 2.5min at 110°C (unexposed part cures);
5. Develop in Toluene for 15sec;
(6) Post bake at 110°C for 2-3 min.

Figure 27 shows the pattern designs (circular-holes and serpentes) and the actual patterned PDMS substrate using this method. Analytical calculation shows the circular-hole pattern with a radius-to-spacing ratio of 30% can reduce the effective modulus of the PDMS substrate to as small as 10% of its original value.

![Photo-patternable PDMS design and actual samples](image)

**Figure 27.** Photo-patternable PDMS design and actual samples

Future work of this ongoing project includes conducting mechanical tests to measure the actual effective modulus and comparing with the calculation, and applying this method to substrates with three-dimensional geometries (such as the finger tube).

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7 Conducted by Y. Su, R. Li and H. Cheng, Department of Mechanical Engineering and Department of Civil and Environmental Engineering, Northwestern University.