Supplementary Material for

Soft Microfluidic Assemblies of Sensors, Circuits, and Radios for the Skin

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**Materials and Methods**

**Design rationale for the circuits**

For the inductive electrocardiogram (ECG) system, Chip 105 (10 µF capacitor) is used to block the DC offset of the incoming ECG signal. Chip 107 (0.1 µF capacitor) and chip 13 (200 kΩ) form a high pass resistor-capacitor filter to remove low frequency noise from the incoming ECG signal. The instrumentation amplifier (AD627b, chip 1) amplifies the ECG signal by 800 times. Chip 107 (0.1 µF capacitor) and chip 10 (10 kΩ) form a low pass resistor-capacitor filter to remove high frequency noise from the amplified ECG signal collected by the epidermal working and reference electrodes, with a floating ground. A voltage controlled oscillator (MAX2750, chip 3) generates high frequency signals at ~2.4 GHz that represent the ECG signal voltage. The output of this chip passes out of the device through a compact three dimensional antenna (chip 18). A separate patch antenna, radiofrequency (RF) amplifier and frequency counter can receive the transmitted RF wave reliably at a distance of up to 1 m, when operated in a room designed to eliminate background electrical noise. The inductive power module is based on resonant inductive coupling: a 10 µH inductive coil (27T103C, chip 101) and 0.1uF capacitor (chip 107) forms an inductor-capacitor loop. A Schottky diode provides rectification and a capacitor (10 µF) provides integration.

For the multifunctional device, the electrophysiological (EP) amplification subunit is similar to that of the ECG system. The acceleration sensing module consists of chip 6 (KXTH9) which converts acceleration to an analog voltage output. The temperature sensing module uses a Wheatstone bridge circuit to convert temperature to an analog voltage output. Chip 5 (PTS080501B500RP100) is a resistance temperature detector. Chips 16 and 17 are paired resistors to form the bridge loop. In the multiplexing module, chip 21 (LTC6991) generates an oscillating signal that controls the multiplexer (MAX4734, chip 2), to switch sequentially among the outputs of the three sensor channels, EP, temperature and acceleration sensing, to allow data transmission with a single voltage controlled oscillator. In particular, data from the EP, acceleration and temperature channels transmit for 18, 9 and 9 seconds, respectively, as determined by two oscillators that generate two bit control logic by tuning the ratio of their oscillation frequencies to 1:2. A pulse width modulation oscillator periodically turns on and off the voltage controlled oscillator, for a duty cycle of ~7% at a frequency of 1000 Hz, to reduce the average power consumption from ~40 mW (continuous operating mode) to ~6.3 mW. The wireless data transmission subunit is similar to that of the ECG system.

**Molding structures of surface relief on the elastomeric device substrate**

The process began with a piece of a clean (100) Si wafer, coated with a 100 nm thick Si₃N₄ film formed by plasma enhanced chemical vapor deposition (STS PECVD). Photolithography (AZ P4620, 3000 rpm, 30 s, soft bake at 110 °C for 3 min, 300 mL/cm², 1:2 volume ratio of AZ 400K and de-ionized water for 1 min) and reactive ion etching (RIE, 22.5 sccm CF₄, 40 mT, 150 W, 8 min) defined an array of circles (20 µm) in the Si₃N₄. Wet chemical anisotropic etching of the silicon with KOH (100 ml 33% weight percent water solution, 20 ml isopropanol alcohol, 130 °C, 45 min, with strong magnetic stirring) generated an array of pyramidal shaped recessed regions on the surface of the wafer. After removing the remaining Si₃N₄, a second photolithography step (AZ P4620)
and inductively coupled plasma reactive ion etching process (ICP RIE, Bosch process) defined trenches (30 µm in depth) for the support posts for the chips. A third photolithography step (AZ P4620) and RIE process (STS ICP RIE, Bosch process) defined the trenches (100 µm in depth) for the isolation barriers. A layer of polytetrafluoroethylene (~200 nm) conformally deposited (STS ICP RIE) all exposed surfaces to minimize adhesion. Thin (300 µm) silicone substrates (Ecoflex, Smooth-On) were prepared by mixing the two components of a commercial kit in a 1:1 weight ratio, spin-casting (300 rpm for 30 s) the resulting material onto the processed Si wafer and then curing into a solid form (2 hours at room temperature).

**Fabrication of the interconnect network**

The process began with spin casting of polydimethylsiloxane (PDMS, Sylgard 184), mixed at 10:1 ratio, onto a clean glass slide at 3000 rpm for 30 s. After curing in an oven at 70 °C for 2 hours, the PDMS was exposed to oxygen plasma (20 sccm O₂, 300 mT, 200 W, 30 s). A 2.4 µm thick layer of PI (from poly(pyromellitic dianhydride-co-4,4'-oxydianiline) amic acid solution was then applied by spin casting (2000 rpm for 60 s), baked on a hotplate at 150 °C for 4 mins and in a vacuum oven at 10 mT and 250 °C for 1 h. The interconnects and metal electrodes consisted of a 400 nm thick layer of Cu deposited by electron beam evaporation onto the PI. Photolithography (AZ P4620) and etching (CE-100 copper etchant, Transene Company) defined patterns in the Cu. Next, spin coating formed a second 2.4 µm thick layer of PI over the entire structure. A 50 nm thick layer of SiO₂ was then deposited using electron beam evaporation, to serve as an etching mask for the PI. Next, photolithography (AZ P4620), RIE etching (50 mT, 40 sccm CF₄, 100 W, 20 min), and oxygen plasma etching (20 sccm O₂, 300 mT, 200 W for 21 mins) patterned the layers of PI in a geometry matched to the metal traces. The residue SiO₂ mask was removed using buffered oxide etchant, and the overall circuit electrodes were immersed in electroless Sn plating solution (Transene Company) at 80 °C for ~10 s. The Sn deposited only onto the exposed Cu surfaces, for the purpose of ensuring good wettability of the solder on the bonding pads. Finally, the circuit electrodes were retrieved using water soluble tape (3M, Inc.) for aligned transfer to the device substrate.

**Assembly of the chip components**

Electron beam evaporation of Ti (5 nm) / SiO₂ (50 nm) on the pads (mounting sites for the chips) of the interconnect network formed backside coatings (30). A shadow mask made of PI (Stencilunlimited corp.) with patterns matched to the support posts, was aligned and laminated onto the molded silicone substrate (thickness ~0.4 mm). The substrate was then activated by exposure to ultraviolet induced ozone for 5 mins. Aligning and laminating the interconnect network onto this surface led to an irreversible strong bonding upon contact, only at the locations of the support posts (~1 mm diameter). After aging for 10 mins, the water soluble tape was removed by immersion in tap water for 1 hour. Another shadow mask, the same as the one for electroless Sn plating, was then aligned and laminated onto the electrodes to selectively expose the contact pads for the pin contacts associated with each of the component chips. A Sn₄₂Bi₅₈ alloy solder paste (Chip Quik Inc. SMDLTFP250T3) was screen printed onto the contact pads. A combined lapping and polishing process with tripod polisher was used for thinning of the thickest
chips, such that all chips had thicknesses <1 mm. The lapping process involved a fast back grinding step, followed by a polishing process to remove remaining rough surfaces. Each chip component was manually placed over the electrodes under an optical microscope. After all of the chips were in position, the solder paste was reflowed in an oven at ~180 °C for 5 mins. Good solder joints appeared smooth and shiny, with complete wetting to the contact pads. A thin silicone superstrate (~100 µm thick) encapsulated the entire chip region of the device. The edges were sealed with an additional application of partially cured silicone followed by baking on a hotplate at 120 °C for 10 min. A liquid PDMS base (Sylgard 184, without the curing agent), injected into the capped cavity through an edge, covered all of the chips and interconnects via capillary force.

Mechanical testing and simulation of the devices

An array of metal dots (100 nm Cr, 0.4 mmØ for each dot, 1 mm pitch) was deposited through a polyimide shadow mask mounted on the back side of the relief substrate. After integrating the interconnect network and chips and adding a thin layer of PDMS base, equal-biaxial stretching was applied to the device using a customized stage. Strain was added/removed gradually and simultaneously in both directions. Images of the device at various stages of deformation were collected with a digital single-lens reflex camera from the backside the device so that the metal dots were clearly visible. The Young’s moduli of the devices with and without the chips and interconnect network were measured in orthogonal directions using an INSTRON MINI44. The strain-stress curves were averaged over at least three individual measurements. Mechanical simulation was performed using finite element analysis (FEA) techniques.

Full three-dimensional (3D) FEA was adopted to analyze the postbuckling behaviors of the entire device under uniaxial and biaxial stretching. The chips were selectively bonded to the silicone substrate (Ecoflex; thickness 0.5 mm) via small circular (diameter 1 mm) and rectangular (0.5 mm by 1.0 mm) pedestals. Each of the metal interconnect (Cu, thickness 400 nm) lines, was encased, top and bottom, by a thin layer of polyimide (PI, thickness 2.4 µm for each layer). The elastic modulus (E) and Poisson’s ratio (ν) are \( E_{\text{Ecoflex}} = 0.0623 \text{ MPa} \) and \( v_{\text{Ecoflex}} = 0.49 \) for Ecoflex; \( E_{\text{Cu}} = 119 \text{ GPa} \) and \( v_{\text{Cu}} = 0.34 \) for copper; and \( E_{\text{PI}} = 2.5 \text{ GPa} \) and \( v_{\text{PI}} = 0.34 \) for PI. Eight-node 3D solid elements and four-node shell elements were used for the ecoflex and self-similar electrode, respectively, and refined meshes were adopted to ensure the accuracy. Linear buckling analyses were carried out to determine the critical buckling strain and lowest buckling mode for each interconnect, which were then implemented as initial geometric imperfections in the postbuckling simulation. The evolution of deformed configurations with applied strains were obtained from FEA for the entire device under both uniaxial and biaxial stretchings, as shown in Figs. 2, S11, S13, and S14. Good agreement between FEA and experiment results can be found.

Functional testing of the devices on human subjects

All experiments on human were conducted under approval from Institutional Review Board at the University of Illinois at Urbana-Champaign (protocol number: 13398). There are three subjects (age: 21–29, all males), co-authors in the paper. Research was carried out with informed signed consents from the subjects. Prior to
device integration, hairs were removed and the skin was cleaned using a mild abrasive and Scotch tape to exfoliate some of the stratum corneum and to remove sebaceous oils (31). Afterwards, the device was placed on the desired areas on the skin for recording of EP signals. For near field coupling powering, a high frequency alternating current source (10 V_{pp} at 150 kHz) to the primary coil was generated using a KEITHLEY 3390 50 MHz arbitrary waveform generator. The input and output characteristics of the wireless coil were measured using an Agilent infinium DSO8104A oscilloscope (1 GHz, 4 channels). The ECG data were received by a back end antenna, then filtered through a 2.2-2.7 GHz band pass filter (RF-lambda, RBPF2450, SN: 12041902227), amplified by a 2.2-2.7 GHz low noise amplifier (ZQL-2700MLNW+), and read out by a frequency counter, where the frequency-modulated signals were processed and analyzed. For recordings with commercial equipment, we used a pair of Au/Ag ring electrodes, fixed with an adhesive sticker to the skin along with conductive electrode gel. A ground electrode was attached on a more proximal section of the right arm. The wireless system, placed directly adjacent to the electrodes used for measurement with the commercial system (Brain Vision V-Amp, with a nearby ground electrode on a distal position on the left forearm), utilized a floating ground. Recorded data were passed through band pass filters to remove line noise and other high frequency artifacts, as well as slow drifts in the signal. The mean voltage of each signal was subtracted to remove the DC offsets. The data were then normalized to their peak values, to facilitate comparisons.
Fig. S1.
Optical image of the non-coring needle, highlighting the inject pore on the side wall of the needle. This construction avoids removal of subject material during the injection process. The elastomer self-seals after removal of the needle. Additional elastomer can be added to enhance the robustness of this seal.
Fig. S2
Thermogravimetric analysis of 10.12 mg PDMS base fluid at elevated temperatures for prolonged time.
Fig. S3

3D-FEA results that illustrate the mechanics advantage of free-floating interconnects in comparison with fully bonded or embedded ones. (A) Initial configuration of one of the interconnects shown in Fig. 1C; (B) the strain distribution of the interconnect only with two ends bonded onto the top of the Ecoflex, when the applied strain reaches the corresponding elastic stretchability (167%) of the interconnect; (C) the strain distribution of the interconnect fully bonded onto the top of the Ecoflex, when the applied strain reaches its elastic stretchability (17%); (D) the strain distribution of the interconnect fully embedded in the middle of the Ecoflex, when the applied strain reaches its elastic stretchability (8%).
**Fig. S4**

**Characterization of the hierarchical substrate.** (A) Optical image of a substrate with multiple levels of embossed relief. (B) SEM image of a small region of the substrate to illustrate its hierarchical structure.
Fig. S5
X-ray side view image of a typical commercial chip with standard package design. The thickness of this chip can be reduced to 0.86 mm by grinding the excessive packaging epoxy away.
Optical images of the interface wetting behavior of the interconnects. Chips bonded on the interconnects (A) before and (B) after electroless Sn plating. These images clearly show that after Sn plating, the $\text{Sn}_{42}\text{Bi}_{58}$ solder paste, applied by screen printing, can wet the interconnector surfaces well and thus allow a robust bonding interface.
Fig. S7
Layout design of the single ECG system, with labels for different parts of the device. The design of the transition region between the interconnectors and the chip bonding pads, to minimize stress concentration, is also highlighted in the dashed purple box.
Comparison of mechanical performance of three different interconnect designs at the point of the periphery of the microfluidic enclosure. (A) Schematic illustration (left panel) of a self-similar serpentine interconnect across the interface, and the distribution of maximum principal strain (middle panel for horizontal stretching along the x direction; right panel for vertical stretching along the y direction) in the metal layer for 100% uniaxial stretching. (B) Schematic illustration (left panel) of a straight interconnect (length 1 mm) across the interface, and the distribution of maximum principal strain (middle panel for horizontal stretching; right panel for vertical stretching) in the metal layer for 100% uniaxial stretching. (C) Schematic illustration (left panel) of a straight interconnect (length 0.5 mm) across the interface, and the distribution of maximum principal strain (middle panel for horizontal stretching; right panel for vertical stretching) in the metal layer for 100% uniaxial stretching. The simulation shows that the optimized straight interconnect at the interface avoids fracture inducing strains even under 100% biaxial stretching, in agreement with the FEA results, where the strain of interconnect remains well below the fracture limit (~5%).
Fig. S9
Illustration of the fully integrated device. (A) Schematic illustration of a device in a slightly deformed configuration, and (B) optical image of an actual device in a similar state of deformation, with labels for different parts and modules in the circuit.
Fig. S10
Schematic layout of the ECG system with the chip positions and chip types (Table S1) labeled. Black dots indicate the number 1 pin of each individual chip.
**Fig. S11**

Experimental and computational studies of localized deformations in various self-similar serpentine interconnects within the ECG system, under biaxial stretching. (A) to (J) show optical images and corresponding FEA results for ten self-similar serpentine interconnects as the entire circuit is biaxially stretched from 0% to 50% and 100%. The color in the FEA results represents the maximum principal strains of the metal layer.
Fig. S12
Dependence of the strain in the interconnect metal (at the material level) on the applied strain (at the system level). The maximum value ($\varepsilon_{\text{max}}$) of the principal strain in the metal layer of the interconnect network as a function of the biaxial applied strain ($\varepsilon_{\text{appl}}$), together with illustrations of the evolution of the deformations.
Fig. S13
Experimental and computational studies of buckling deformations across the entire circuits under uniaxial stretching along the horizontal direction. 3D-FEA results (A) and corresponding optical images (B) of the entire circuit when uniaxially stretched from 0% to 10%, 20%, 30% and 40%. The color in the FEA results represents the maximum principal strains in the metal layer.
Fig. S14

Experimental and computational studies of localized deformations of self-similar serpentine interconnects with the circuits under uniaxial stretching along the horizontal direction. (A) to (E) show optical images and corresponding 3D-FEA results of five self-similar serpentine interconnects as the entire circuit is uniaxially stretched from 0% to 20% and 40%. The color in the FEA results represents the maximum principal strains of the metal layer.
Fig. S15
Cycling testing the ECG device under uniaxial strain with amplitude of 30% at a frequency of 0.6 Hz. (A-N) Images of the device at 500 cycle increments, up to 6500 cycles.
Finite element simulations to illustrate the effects of rigid devices on the strain distribution across the soft substrate under biaxial stretching of $\varepsilon_{\text{appl}}=20\%$. (A) The distribution of logarithmic strain in the substrate when the rigid devices are selectively bonded to the substrate (via small circular and rectangular pedestals). (B) The distribution of logarithmic strain in the substrate when the full regions of each device are bonded to the substrate.
Fig. S17
Strain mapping at different levels of strain when all of the chips are fully bonded on the substrate. The strain localization can be clearly visualized under these chips.
3D-FEA of epidermal electrodes with self-similar serpentine mesh designs. (A) A traditional serpentine mesh design with a filling ratio of ~31%, and the strain distribution when the applied strain reaches the elastic stretchability (12.5%). (B) The self-similar serpentine mesh design with a filling ratio of ~31%, and the strain distribution when the applied strain reaches the elastic stretchability (25.0%). Both the top view and angled views of the deformed configurations are shown in the figure, indicating smaller wrinkling wavelength for the self-similar serpentine mesh design on the right.
**Fig. S19**

**Illustration of the experimental setup and measurement position on the human body.** (A) Experimental setup to measure the ECG in a room design to eliminate background electrical noise. (B) Different lamination positions on the human body for measuring the electrophysiological signals.
Fig. S20

Comparison of the ECG signals from this work and commercial systems. (A) ECG data acquired using a device mounted on the sternum (green) and simultaneous measurement using a wired commercial device with commercial electrodes (blue) placed next to those of the wireless system. (B) The expanded graph provides a detailed view that shows the expected QRS complex. The data are plotted in arbitrary units, normalized and offset in the vertical direction to facilitate comparisons.
**Fig. S21**

A time-frequency spectrogram of the EEG computed with wavelet decomposition. EEG data from Fig. 3F, in normalized arbitrary units were decomposed using a 20 cycle wavelet transform. During mental math, an increase in high-frequency activity between 12-40 Hz (Beta band) is observed that dissipates with rest. During rest, the data show a clear peak in the lower frequency delta band that is larger than that during math.
Fig. S22
EOG and EMG data with commercial device comparison. (A) EOG recorded at positions superior and inferior to the left eye (and slightly lateral from the left outer canthus) as the subject blinks once per second (green), and simultaneous measurement using a wired commercial device with commercial electrodes (blue) placed next to those of the wireless system. The right plot shows data during the time of two eye blinks. (B) EMG measured from the proximal left forearm over the flexor carpi radialis muscle during hand clenches (green), and simultaneous measurement using a wired commercial device with commercial electrodes (blue) placed next to those of the wireless system. The data from the wireless device were inverted along the vertical axis, to facilitate comparison. The right plot shows data corresponding to a single muscle contraction. The commercial system used a pair of Au plated Ag electrodes held against the head with adhesive stickers and conductive electrode gel, and a nearby ground electrode.
Fig. S23
Working principle of the multiplexer.
Fig. S24

The multifunctional sensor system with different parts and chip information (Table S3) labeled. Black dots indicate the number 1 pin of each individual chip.
Design strategy for reducing the power consumption of the voltage controlled oscillator (VCO), by adding a pulse modulator. When the VCO duty cycle is reduced to 7%, the power consumption of the entire system is about 6 mW.
Releasable power supplies for the multifunctional system. (A) Optical image of the NFC module from the front and back sides, and associated chip information (Table S1). (B) Optical image of the front and back sides of a small coin cell battery module. Both of these modules can laminate onto the power supply contact pads of the multifunctional device, to provide power for operation.
Fig. S27
Images of the device on the skin. (A) Top down view to show its overall footprint (1.9 x 2.9 cm²), and (B) perspective view to show its overall thickness (~1 mm) with reference to a coin (US quarter).
Fig. S28
Calibration curve for the temperature-frequency relationship in the multifunctional system. The error bars for each data point are from three independent measurements.
Fig. S29

Signal readouts from the multifunctional device when mounted on the forearm during periodic clinching and relaxing of the fist. (A) Acceleration signal and (B) temperature signal before and after 9 min of clinching and relaxing. (C) The temperature measured before (left panel) and after (right panel) this exercise, as confirmed by the commercial IR camera.
Table S1
Chip information of the ECG system.

<table>
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<th>Type</th>
<th>Value</th>
<th>Manufacturer part number</th>
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<td>Jumper</td>
<td>0 Ω</td>
<td>C80406000000200BAHP</td>
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<td>1</td>
<td>Operational amplifier</td>
<td>N/A</td>
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<td>3</td>
<td>Voltage controlled oscillator</td>
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<td>4</td>
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<td>Inductor</td>
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Table S2

**Computational model for the effective Young’s modulus.** The moduli for the experimental data are determined by using linear fits of the stress-strain curve in the range of [0%, 50%].

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<th>FEA</th>
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<th>Modulus (x)</th>
<th>Modulus (y)</th>
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<td>With chip</td>
<td>172.06 kPa</td>
<td>193.80 kPa</td>
<td>171.20 kPa</td>
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<td>Without chip</td>
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<td>Average</td>
<td>3.2%</td>
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<td>5.5%</td>
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**Table S3**  
Chip information for the multifunctional system.

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<th>Manufacturer part number</th>
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<td>IC multiplexer</td>
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<td>3</td>
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Movie S1

Video of the self-alignment of chips during the soldering/bonding process. Under controlled heating, the solder paste reflows and wets the serpentine metal electrodes and the contact pads on the chips. The surface tension causes motion that maximizes the contact area to the metal electrodes underneath, thereby self-aligning the chips.
Movie S2

This video shows an ECG device during operation. This device has three basic functions: wireless power transfer, electrophysiological potential sensing and wireless data transmission to an external receiving system. When the inductive coil is placed near the secondary coil in the device, the ECG signal from the chest of a test subject was amplified, filtered and wirelessly transmitted to a receiver. The resulting data appears in real time on the computer screen, as it is recorded. The drifting baseline of the signal in the first few seconds is because of a capacitor connected in series in the circuit.
References and Notes


25. Y. Lee, S. Bang, I. Lee, Y. Kim, G. Kim, M. H. Ghaed, P. Pannuto, P. Dutta, D. Sylvester, D. Blaauw, A modular 1 mm$^3$ die-stacked sensing platform with low power $^1$C inter-die


