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Wireless batteryless soft sensors for ambulatory cardiovascular health monitoring

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Abstract

Seismocardiography (SCG) is the measure of local vibrations in the chest due to heartbeats. Typically, SCG signals are measured using rigid integrated circuit (IC) accelerometers and bulky electronics. However, as alternatives, recent studies of emerging flexible sensors show promise. Here, we introduce the development of wireless soft capacitive sensors that require no battery or rigid IC components for measuring SCG signals for cardiovascular health monitoring. Both the capacitive and inductive components of the circuit are patterned with laser micromachining of a polyimide-coated copper and are encapsulated with an elastomer. The wearable soft sensor can detect small strain changes on the skin, which is wirelessly measured by examining the power reflected from the antenna at a stimulating frequency. The performance of the device is verified by comparing the fiducial points to SCG measured by a commercial accelerometer and electromyograms from a commercial electrode. Overall, the human subject study demonstrates that the fiducial points are consistent with data from commercial devices, showing the potential of the ultrathin soft sensors for ambulatory cardiovascular monitoring without bulky electronics and rigid components.

Keywords: Soft sensor, capacitive strain sensor, passive wireless, laser micromachining, LC resonance sensor, ambulatory monitoring, cardiovascular health

INTRODUCTION

Cardiovascular diseases (CVD) account for over 19 million deaths annually in the United States[1]. Early detection and diagnosis of this broad group of diseases are imperative in treatment. The clinical standard for detecting cardiovascular abnormalities is the electrocardiogram (ECG)[2]. However, ECG alone cannot capture a holistic view of heart function[3]. Therefore, numerous other cardiovascular signals have offered crucial information to assist in the detection of CVD[4]. One of these signals is the seismocardiogram (SCG). SCG is traditionally measured using an accelerometer to capture the local vibrations of the heart[3]. While ECG measures myocardial conduction, SCG measures myocardial contractions and can capture the timing of fiducial points, such as mitral valve openings (MO) and closings and atrial valve openings and closings, which ECG cannot capture[3,5]. SCG is typically measured using integrated circuit (IC) accelerometers[3]. However, these accelerometer systems are rigid and bulky, making them more uncomfortable to wear for long periods. In addition, accelerometers are also prone to the global inertia of the body[6]. Although thin and flexible electronics are now becoming a widespread research field, exploring soft sensors for SCG is relatively unexplored. Soft and flexible sensors can offer solutions to these problems[7-12]. Liu *et al.* demonstrated a flexible serpentine interconnecting device over a rigid printed circuit board[5]. However, the fabrication was still largely dependent on cleanroom processes and photolithography.

Ha *et al.* first reported a stretchable piezoelectric strain sensor based on polyvinylidene fluoride (PVDF) that could stretch 110%[7]. Nayeem *et al.* also reported a PVDF-based mechanoacoustic sensor with high sensitivity[13]. Additionally, Lo Presti *et al.* reported an SCG sensor using optical strain sensors[14]. All of these are improvements on the traditional MEMS accelerometers, yet they still require additional components for transmission. Adding the seamless wireless capability to these thin sensors is essential for ambulatory monitoring and patient comfort. Standard accelerometer-based SCG sensors have communicated data using Bluetooth[15], WiFi[16], and Zigbee[17]. On the other hand, thin-film SCG sensors have only been measured wirelessly using active near-field communication (NFC) components[18]. Additionally, these sensors are still limited by rigid circuit components and power consumption limits[18]. All the current SCG sensors are either limited by low-throughput fabrication processes or bulky hardware for transmission. Thus, in this work, we aim to combine the comfort of thin film SCG sensors, a rapid one-step fabrication method, and passive wireless capability.

Here, we propose a batteryless, wireless, chipless, and passive soft SCG sensor that can be measured using inductive coupling. The sensor and coil are incorporated in a single layer of copper, backed by polyimide (PI), eliminating the need for cleanroom processes or other time-consuming methods. The thin sensor is encapsulated with elastomer and attached to the skin using a low-modulus silicone gel, allowing the device to attach to the human skin with limited restrictions conformally. Furthermore, the passive sensor does not require any batteries or IC components, allowing the entire sensor to be flexible, stretchable, and nearly invisible to the user. To validate the sensor, both an SCG measured with a commercial accelerometer and a single-lead ECG were used to compare the fiducial points. The fiducial points, including mitral valve closing (MC), isovolumic movement (IM), aortic valve opening (AO), aortic valve closing (AC), and MO[19], showed a strong association between the reported device and the commercial accelerometer.

EXPERIMENTAL

Assembly of passive sensors

First, a glass slide was spin-coated with silicone elastomer (Ecoflex 00-30, Smooth-On, Inc.) at 500 rpm for 30 s and cured in an oven at 60 °C for 20 min. On a separate glass slide, polydimethylsiloxane (Sylgard 184,

Sylgard) was spin-coated at 800 rpm for 60 s. Copper foil (6 μm, BR0214, MSE Supplies LLC) was then laid across the slide. PI (PI2545, HD MicroSystems) was spin-coated onto the copper foil at 800 rpm for 60 s and cured on a hot plate at 240 °C for 60 min. Afterward, the copper and PI were transferred to the glass slide coated with elastomer. The sensor was then patterned on a laser micro-machining system (Femtosecond Laser Micro-Machining System, OPTEC). A bridge made of commercial high gauge enameled magnet wire was soldered across the pads connecting the interior of the coil with the end of the capacitive sensor. Elastomer was spin-coated across the device at 500 rpm for 30 s, and the device outline was cut out using laser micromachining. Finally, a thin coat of high tack silicone gel (A-4717, Factor 2) was added to one side of the sensor for adhesion to the skin. The assembly steps are shown in Supplementary Figure 1.

Wireless transmission

The capacitive sensor and coil form an LC circuit with a varying capacitive value. When the capacitance of the sensor decreases, the resonant frequency of the circuit increases. A vector network analyzer (Tektronix TTR506A) was connected to a spiral receiving antenna. A custom MATLAB program was used to record the S_{11} parameter at a stimulus frequency close to the resonant frequency. The resonant frequency was determined by locating the frequency at the minimum S_{11} value in the range of 10-60 MHz. The recording stimulus frequency was then determined by subtracting 0.1 MHz from the resonant frequency.

Validation

Commercial Ag/AgCl electrodes across Lead I were used for the reading of the ECG, and a low-power commercial accelerometer (ADXL355, Analog Devices, Inc.) was used for the reference SCG.

RESULTS AND DISCUSSION

We developed a wireless, ultrathin, and soft sensor with no rigid components such as batteries and IC chips, demonstrating a seamless integration with the human skin. When mounted on the chest, this device can measure high-fidelity strain changes due to cardiac vibrations. Figure 1A shows the mounting location of the wearable sensor when placed on the lower chest. Figure 1B captures a series of photos showing a very small form factor and mechanical compliance of the thin-film soft sensor. This device, encapsulated with elastomeric membranes, shows superior stretchability and flexibility with multi-modal bending, pulling, and twisting. Figure 1C shows a schematic of the measurement system of the device and its capabilities in passive wireless sensing using antennas. The device comprises a capacitive strain sensor and an inductive coil, forming an LC oscillator circuit[20,21]. This passive circuit offers wireless detection of the measured capacitive signals without rigid circuits and batteries. When mounted on the skin, this sensor is inductively coupled with an external antenna to record strain changes caused by cardiac vibrations. Overall, the soft sensor can measure SCG, pulse, and heart rate data. Table 1 captures the unique advantages of the developed sensor when compared to the prior work showing different types of sensors.

Figure 2A shows the layer structure of the developed chip-less, battery-less sensor. The thickness of the entire device is 471 μm. The fabrication of the sensor only requires the patterning of one layer. Using a femtosecond laser, the sensor is fabricated by micromachining a 6-μm copper foil backed by PI. Laser micromachining was chosen as the fabrication process since it represents a promising alternative to traditional cleanroom lithography processes. To avoid direct contact with the human skin, the sensor is encapsulated in a low-durometer silicone elastomer, providing conformal contact with the skin. An additional layer of high tack silicone gel was added between the device and the skin to maximize conformality[2]. It was also important for the elastomer to be as thin as possible to maximize conformality to the skin. Since copper is not stretchable, the fingers are supported by connections with serpentine structures [Figure 2B]. The laser micromachining process allowed interdigitated electrode spacings to be as small as

Table 1. Comparison of flexible SCG sensors

Reference	Sensor type	Material	Transmission	Battery-free	Sensitivity
This work	Capacitive strain sensor	Copper	LC resonance coupling	Yes	2 MHz per 1% ε
[7]	Piezoelectric strain sensor	PVDF	Wired	No	0.4 mV per με
[13]	Piezoelectric strain sensor	PVDF nanofiber	Wired	No	10 V per Pa
[18]	Piezoelectric strain sensor	PVDF	NFC	Yes	Not reported
[14]	Fiber-optic strain sensor	Acrylate fiber	Wired	No	0.045 nm per mε
[22]	Resistive strain sensor	Porous graphene	Wired	No	96 Ω per 1% ε

NFC: near-field communication; PVDF: Polyvinylidene fluoride; ε: strain.

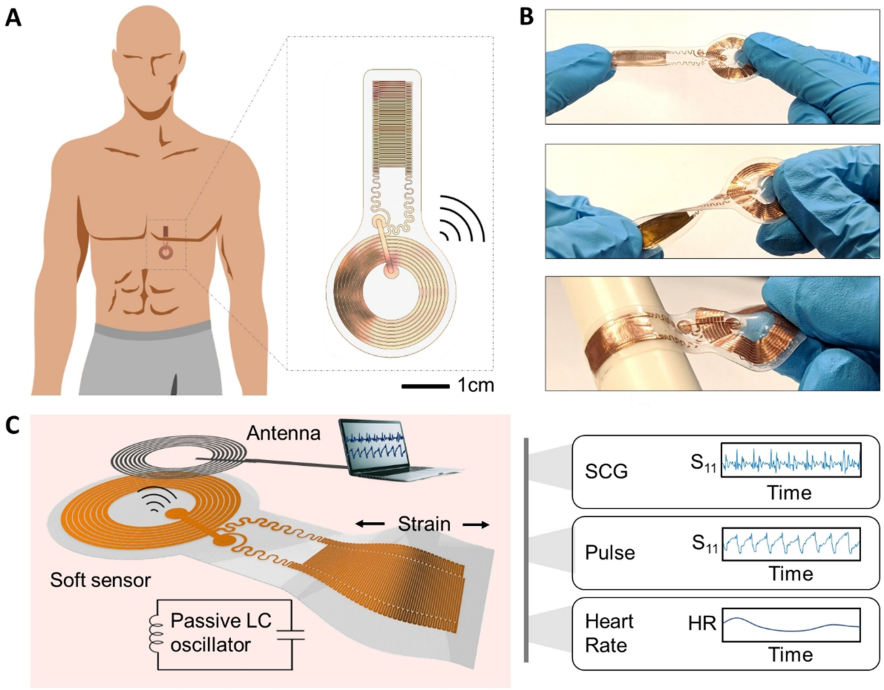

Figure 1. Overview of a wireless soft sensor system for cardiovascular health monitoring. (A) A soft capacitive sensor mounted on the chest for wireless detection of data; (B) Photos of a fabricated thin-film soft sensor showing flexibility and stretchability; (C) Schematic of the inductive coupling measurement method. The soft wearable sensor can measure seismocardiography, pulse, and heart rate. SCG: Seismocardiography.

8 μm. This is important since the spacing between the interdigitated electrodes of the capacitive strain sensor determines the sensitivity[23]. As the sensor is stretched, these serpentine structures expand to allow a large difference in spacing between the fingers. In addition, the serpentine pattern changes between the coil and the sensor to enable strain at the interface while minimizing stress on the copper [Figure 2C]. To show the reliability of the sensor, it was stretched and unstretched for 100 cycles. Figure 2D shows repeatable capacitance measurements with negligible hysteresis as a function of strain. The sensor also shows great stability and reliability; the capacitances stay constant for 100 cycles [Figure 2E]. The sensor shows virtually no degradation over the cycles, as the amplitude of the capacitance of the 100th cycle shows the same amplitude of the 1st cycle (0.362 pF). To measure the relationship between the S_{11} parameter and the strain, an experiment was conducted where continuous frequency sweeps were conducted at varying sensor strains. Figure 2F shows a consistent resonant shift based on the strain. This was important to the design of

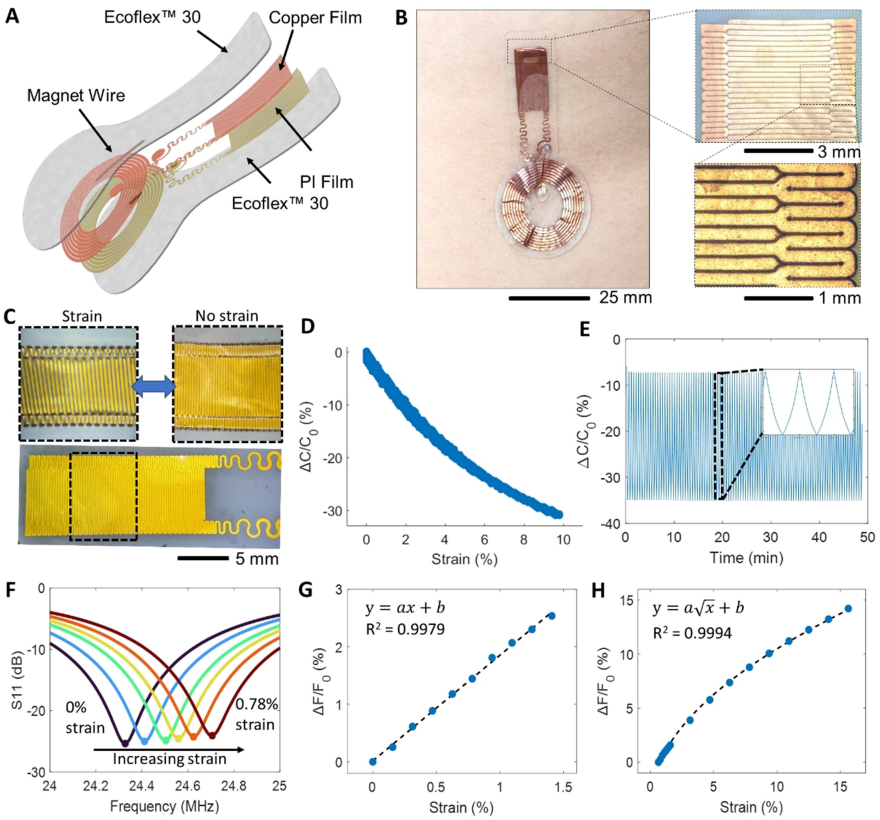

Figure 2. (A) Schematic of the sensor layers; (B) Photograph of the sensor of human skin with a close-up of the fingers; (C) Photograph of the sensor mechanism with and without strain; (D) Sensor capacitance as a function of strain; (E) Capacitance changes of the sensor during 100 stretching cycles; (F) Resonant frequency sweeps at different strains. The resonant frequency increases with increasing sensor strain; (G) Resonant frequency at low strain values; (H) Resonant frequency at high strain values.

the device as the S_{11} parameter is the measurement we use for capturing the SCG signal. The sensor shows an excellent linear resonant frequency response in low strains [Figure 2G]. This is especially important for SCG measurement as the skin strain is on the micrometers scale. The sensor showed repeatable measurements at each strain (n = 4), indicating low hysteresis and high signal repeatability. Even at higher strains, the response is still predictable using a square root curve fit [Figure 2H]. The sensor still behaves predictably at strains much higher than 7%, which is the highest strain under normal conditions[24].

The resonance of the LC circuit is measured using an external coil antenna and vector network analyzer. Figure 3 summarizes the inductive coupling method, system setup, and experimental results. Figure 3A and B demonstrates how the sensor is coupled with a vector network analyzer and a data acquisition system. As the capacitance of the sensor decreases, the resonance frequency increases, as described in Supplementary Note 1. To measure the SCG signal, the S_{11} parameter is measured at a stimulus frequency slightly lower than the resonance frequency of the unstretched device. This is to ensure that the polarity of the signal stays consistent throughout the measurement since the direction of the S_{11} parameter with respect to a changing resonant frequency will depend on whether the stimulus frequency is higher than or lower than the resonant frequency. When the stimulus frequency is lower than the resonant frequency, the S_{11} parameter of the stimulus frequency will increase as the resonance frequency increases. This method allows for fast measurement of the SCG signal. Figure 3C shows the S_{11} value at the resonant frequency. As expected, the signal quality decreases as the distance between the coils increases [Supplementary Figure 2]. The coupling

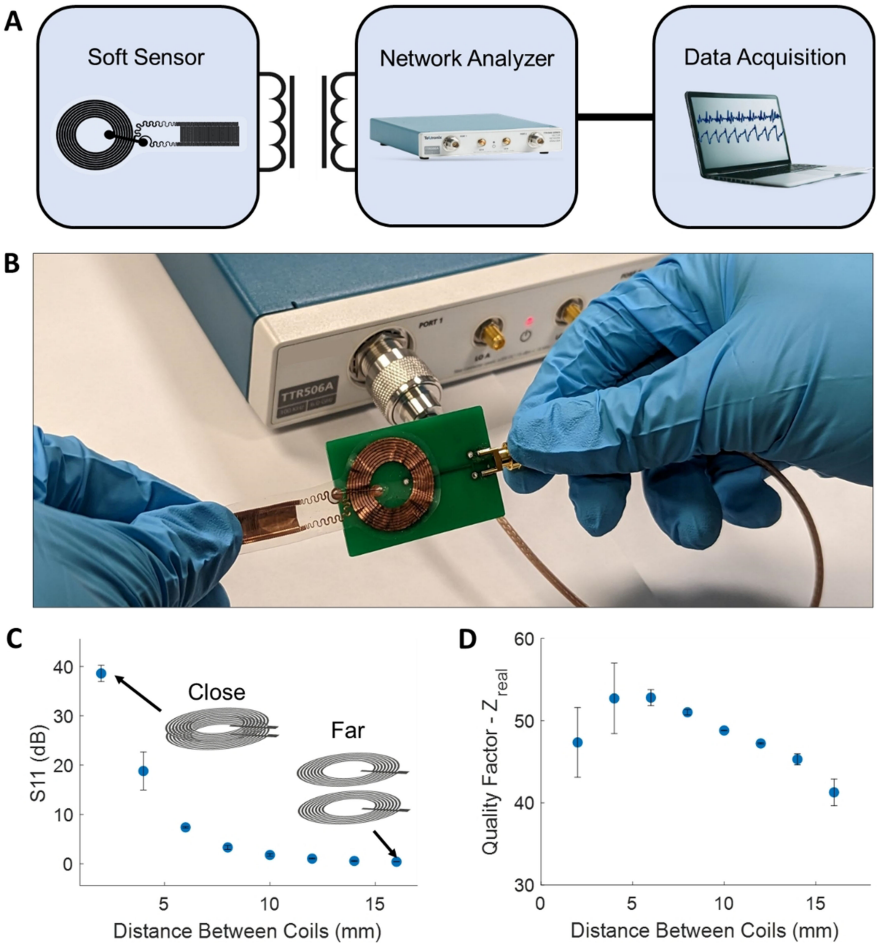

Figure 3. (A) Schematic of the inductive coupling method of detecting signals using the wireless soft sensor; (B) Photo of the inductive coupling system setup. The sensor is attached to the inductive transmitting coil, which is coupled with the receiving coil. The receiving coil is connected to the vector network analyzer; (C) S_{11} at the resonant frequency at different distances between the coils. The coils are kept concentric; (D) Quality factor from resonant frequency sweeps at different distances between the coils. The coils are kept concentric.

shows the highest quality factor at approximately 6 mm between the coils [Figure 3D]. The quality factor is calculated in Supplementary Note 2.

A set of human subject studies have been conducted to validate the performance of our device in detecting SCG signals on the skin. Figure 4 summarizes the experimental results of high-quality detection of heart vibration data. A subject wears the soft sensor patch on the lower chest, as shown in Figure 4A. The ultrathin sensor makes seamless integration and intimate contact with the skin without causing discomfort. In this study, we found that the lower chest location (slightly below and to the right of the left nipple) is the best place to receive large signal amplitudes by the sensor. This location also gives us the advantage of visualizing the pulse rate, which can be seen in the raw data. However, the pulse data are removed in the SCG signal detection by using digital filtering. Figure 4B shows the measured raw SCG signals (top graph) and filtered signals (bottom graph). First, a 1st-order low pass filter at 24 Hz is applied, and a 3rd-order high pass filter at 4 Hz is used. The high pass filter removes baseline wander from the pulse. The low pass filter removes power line noise and any movement artifacts that are not in the desired frequency range, where the desired frequency range for SCG falls below around 24 Hz[25]. In the graph of Figure 4C, the SCG signals are

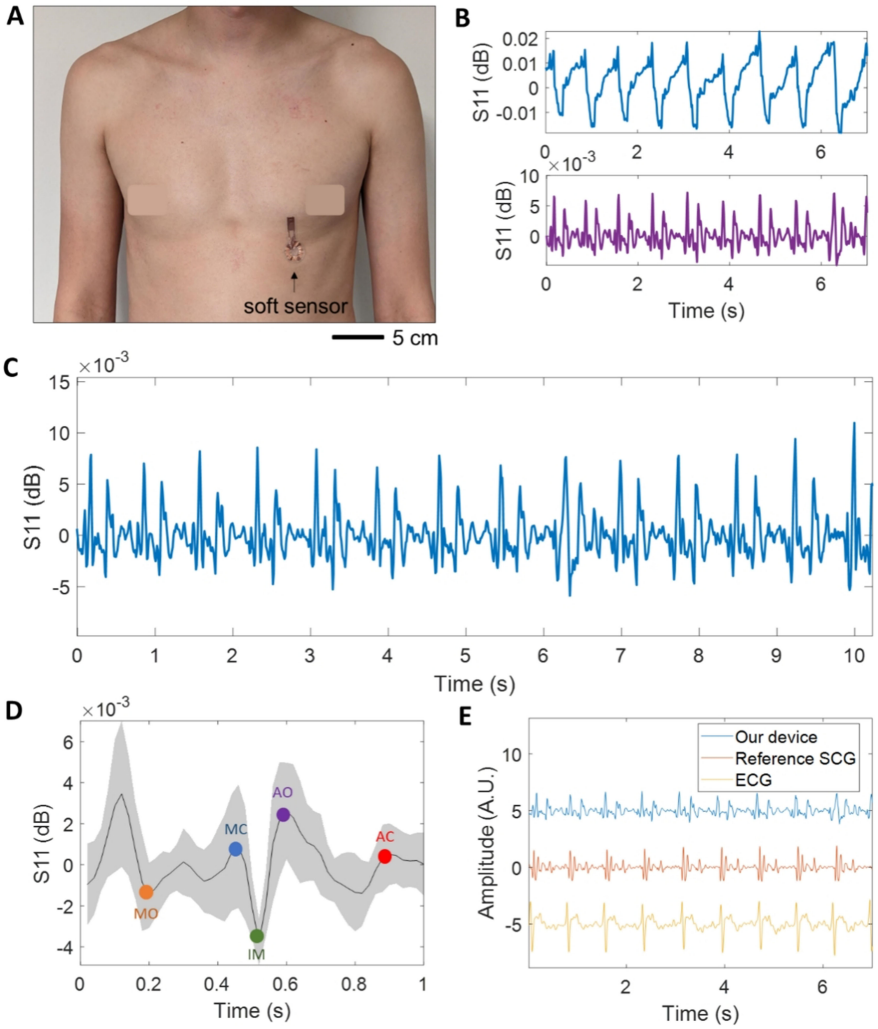

Figure 4. (A) Photograph of the device on the lower chest; (B) Raw seismocardiography (SCG) signals and filtered signals between 4 and 24 Hz; (C) SCG signals recorded during a breath hold; (D) Ensembled averaged SCG signals, showing mitral valve opening (MO), mitral valve closing (MC), isovolumic movement (IM), aortic valve opening (AO), and aortic valve closing (AC) fiducial points; (E) Comparison of SCG data between our device, a commercial SCG device, and a commercial electrocardiogram (ECG) device.

clearly visible for 10 seconds. During this specific measurement, the subjects were instructed to hold their breath to minimize motion artifacts. Tests were also conducted while standing and lying in the supine position [Supplementary Figure 3]. With the system setup, the soft sensor can measure meaningful cardiovascular signals, as demonstrated in Figure 4D, showing the ensemble average using the beats of data and the MO, MC, IM, AO, and AC fiducial points. Figure 4E shows the simultaneous signal measurement of the soft sensor we developed in this work and commercial SCG reference and ECG reference systems [Supplementary Figure 4]. The SCG signals show an excellent agreement on the locations of the S1 and S2 complexes. The soft device also offers the ability to measure systolic time intervals from simultaneous ECG and SCG collection.

CONCLUSIONS

This paper introduces a wireless wearable soft capacitive sensor patch that can measure high-fidelity SCG signals on the skin without using batteries and circuits. The thickness of the entire device is less than 1 mm,

providing an imperceptible lamination to the skin without discomfort. To our knowledge, this device is the first to wirelessly read SCG data accurately without wires or rigid chips. The highly sensitive strain sensor in the patch can measure heart vibrations in the chest and identify fiducial points corresponding to cardiac events using inductive coupling. This stretchable thin-film sensor offers usefulness in ambulatory and remote healthcare applications where mechano-acoustic signals are essential, such as arrhythmia detection or obstructive sleep apnea. Future work will develop more sensors in the patch that would offer high-amplitude signals from the atrial and mitral valves. A dense sensor array can also help to mitigate motion artifacts by eliminating common noise elements. In addition, we will improve the breathability of our device by permeating the substrate and using conductive meshes.

DECLARATIONS

Authors' contributions

Conceptualization, software, formal analysis, investigation, writing - original draft: Guess M
Conceptualization, software, formal analysis, investigation, writing - review and editing: Soltis I, Rigo B
Software, formal analysis, writing - review and editing: Zavanelli N
Formal analysis, investigation: Kapasi S
Writing - review and editing: Kim H
Conceptualization, resources, supervision, project administration, funding acquisition: Yeo WH

Availability of data and materials

The raw data supporting the findings of this study are available from the authors upon reasonable request.

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Conflicts of interest

Yeo WH and Guess M have a patent pending at the Georgia Institute of Technology. Other authors declare no competing financial or non-financial interests.

Ethical approval and consent to participate

Experiments on human subjects were conducted with the approval of the Institutional Review Board of the Georgia Institute of Technology Protocol H22289. All subjects provided informed consent.

Consent for publication

All subjects provided informed consent.

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