Bio-Patch Design and Implementation Based on a Low-Power System-on-Chip and Paper-Based Inkjet Printing Technology

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***Abstract*—This paper presents the prototype implementation of a Bio-Patch using fully integrated low-power system-on-chip (SoC) sensor and paper-based inkjet printing technology. The SoC sensor is featured with programmable gain and bandwidth to accommodate a variety of biosignals. It is fabricated in a 0.18-μm standard CMOS technology, with a total power consumption of 20 μW from a 1.2 V supply. Both the electrodes and interconnections are implemented by printing conductive nanoparticle inks on a flexible photo paper substrate using inkjet printing technology. A Bio-Patch prototype is developed by integrating the SoC sensor, a soft battery, printed electrodes, and interconnections on a photo paper substrate. The Bio-Patch can work alone or operate along with other patches to establish a wired network for synchronous multiple-channel biosignals recording. The measurement results show that electrocardiogram and electromyogram are successfully measured in *in vivo* tests using the implemented Bio-Patch prototype.**

***Index Terms*—Bioelectric system-on-chip (SoC), Bio-Patch, electrocardiogram (ECG), electromyogram (EMG), paper-based inkjet printing technology, sensor IC.**

I. Introduction

THE wide use of multimedia systems and recent advances in solid-state circuit open the door for pervasive and personalized healthcare systems in forms of hospital teleconsultation, remote health monitoring, and ubiquitous mobile telemedicine [1]. Bioelectric signal monitoring can be applied in such multimedia healthcare systems to provide additional valuable information for clinical diagnosis or analysis, especially the vital biosignals monitoring for the elderly health [2], [3]. Electrocardiogram (ECG) and electro myogram (EMG) are the most commonly measured biosignals. Their applications in clinical and research fields have drawn much attention in the recent decade. The authors in [4] presents a telehomecare multimedia platform which not only enables patients video interaction, but also provides the remote online vital sign recording and physiological data measurements, including ECG, heart sound, and blood pressure monitoring.

Generally, nowadays biosignals monitoring is carried out either in a hospital or in ambulatory care. The instruments currently available to the medical communities are either bulky or often limited to one biosignal per device. Several studies attempted to make improvements on the sensing devices or systems for continuous monitoring in daily life. The Holter monitor system is one example. The user wears and records the ECG data for around 24 hours. However, each adhesive electrode requires a cable to connect the human body to the Holter device. As a result, typically six to ten wires are worn around the body. Cumbersome wires incur tangling problem which makes it inconvenient for long term use. Meanwhile, several wearable healthcare systems based on functional textiles are proposed in [5]–[7], using fabric sensing elements to record ECG signals. One drawback is that manual knitting or interconnection processes are involved which makes the production cost high. Some research groups are developing active electrodes using off-the-shelf components with the aim of making them eventually small enough to enable a fully integrated solution [8], [9]. A survey of published works shows that the active electrodes are almost exclusively designed using operational amplifiers. However, active electrodes are not employed as often as expected. The major limitation is that the electronics available to bioengineers are mainly off-the-shelf discrete components. The resulting size becomes a limiting factor when considering integrating complex circuits into electrodes.

A desirable sensor device for personalized and pervasive e-health should be efficient, capable of detecting different biosignals while also able to establish a sensor network for multichannel measurements. It is ideally free of cumbersome cables, and should be small in size, light in weight, low power, low cost, and disposable after use [10]. To realize such a system, principally, two core research challenges are present: low-power electronics and new electrode technologies.

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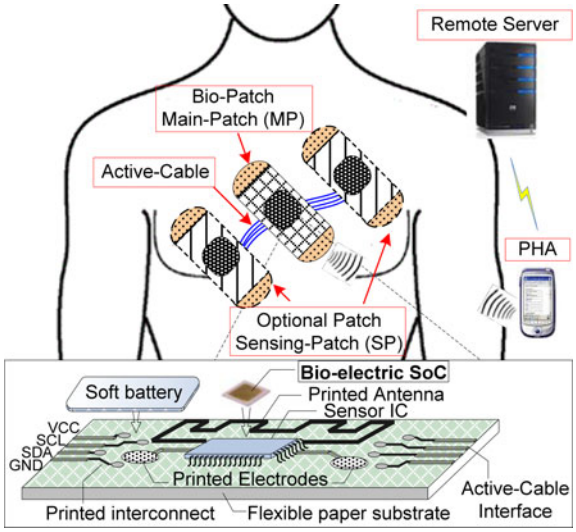

Fig. 1. Concept of Bio-Patch based e-healthcare monitoring system.

Our innovative contribution in this paper is to develop a fully integrated bioelectric system-on-chip (SoC) sensor and integrate it with printed silver electrodes on paper substrate through inkjet printing technology to form a Bio-Patch prototype. The SoC sensor is capable of detecting a variety of biosignals with low power consumption. The inkjet printing technology is a contact-free manufacturing technology. It is employed in this design to print conductive electrodes and interconnections using nanoparticle silver inks on a photo paper substrate. Compared with conventional printed circuit board (PCB) manufacture, it is more cost efficient and environmental friendly, since it does not involve mask processing or etching steps with harsh chemicals.

The concept of the proposed Bio-Patch is illustrated in Fig. 1. It integrates the bioelectric SoC, a soft battery, printed electrodes, and interconnections on a paper substrate. It can be directly attached on human body to collect a specific bioelectric signal. With the SoC sensor, the Bio-Patch is featured with high-quality biosignal extraction by conducting the biosignal amplification, filtering, quantization, buffering, and transmission tasks directly on-site each patch.

A pair of printed electrodes is printed on each Bio-Patch and corresponds to one biosignal channel. The Bio-Patch can work alone or synchronously operate with other patches by sharing the Active-Cable for multichannel biosignals recording. The flat Active-Cable is printed on paper substrate and consists of four silver traces (two for power and ground, and two for serial clock and data: SCL and SDA), over which various commands and digital data are transmitted. If there is only one patch in the system, the patch is configured as a main patch (MP). If there are multiple patches in the system, one is configured as MP and others are configured as sensing patch (SP). The communication between patches is command-based. The MP collects digital biosignals from each SP via Active-Cable. Finally, the collected data are transmitted by the MP to the outside world via wireless link for further applications, such as telehomecare multimedia platform or smart phone based personal health assistant. The system is able to synchronously record up to 14-channel biosignals by using a broadcast command to trigger the acquisition circuit located in each patch startup simultaneously.

With this paper, we report the conceptual design of the Bio-Patch. In particularly, we focus on the prototype implementation and report the preliminary in-lab experiment results that are given in the following sections.

II. Bioelectric SoC

Fully integrated SoC solutions can be found in many advanced medical applications [11]–[13]. The advantages include significant power saving compared with its counterparts using off-the-shelf components. From the perspective of physical size, it makes the sensor miniaturization possible by high level integration: packing the essential signal processing blocks and control units as well as other components on the same chip.

The overall architecture of the implemented bioelectric SoC is illustrated in Fig. 2. The SoC contains three key parts: 1) a three-stage front-end circuit providing a tunable gain and bandwidth, 2) a six-input 8-bit successive approximation register analog to digital converter (SAR ADC), and 3) a digital core. The gain and bandwidth of the front-end can be independently adjusted via external pins to accommodate different biosignals (EMG, ECG, electrooculogram, and electroencephalogram). The amplified biosignal is digitized by the ADC, and buffered in an on-chip RAM. The SoC also includes a built-in two-wire transmission protocol. It is compatible with Philips I^2C interface. The definition of "start-bit" and "stop-bit" follows I^2C specifications, while the broadcast and unicast command packets are custom designed. The serial bus enables the establishment of a body area network over a group of Bio-Patches. The SoC is fabricated in UMC 0.18-μm mixed-mode 1P6M CMOS process and totally occupies 1.5×3 mm^2 silicon area (including IO pads with electrostatic discharge). Preliminary measurement results of the SoC's electric performance are reported in [14]. Fig. 3 shows the unpackaged silicon die on a finger for scale, the microphotograph of the bioelectric SoC, and the measurement PCB for functional tests. A set of switches are employed on the PCB to control the gain and bandwidth of the front-end, input channel selection of the ADC and the configuration of the digital core.

A. Tunable Analog Front-End Circuit

A critical part of the SoC is the analog front-end circuit that defines the quality of the output signal. It consists of a three-stage amplifier chain and a unit buffer, as shown in Fig. 2.

Stage1 is mainly based on a current balanced instrumentation amplifier (CBIA) and a common mode feed back circuit. It provides a fixed gain of 7. A fully differential architecture is adopted in the CBIA to remove input common-mode interference. The output signals of Stage1 are coupled to Stage2 through two capacitors. The close-loop gain of Stage2 is set to 12. In addition, Stage 2 provides a tunable low-pass frequency (LPF) between sub-100 Hz to 1.5 kHz by varying the total capacitance of the on-chip capacitor bank through external switches. Stage3 offers a digitally controlled gain ranging from 2 to 19 by adjusting the equivalent capacitance of an integrated switched capacitor bank.

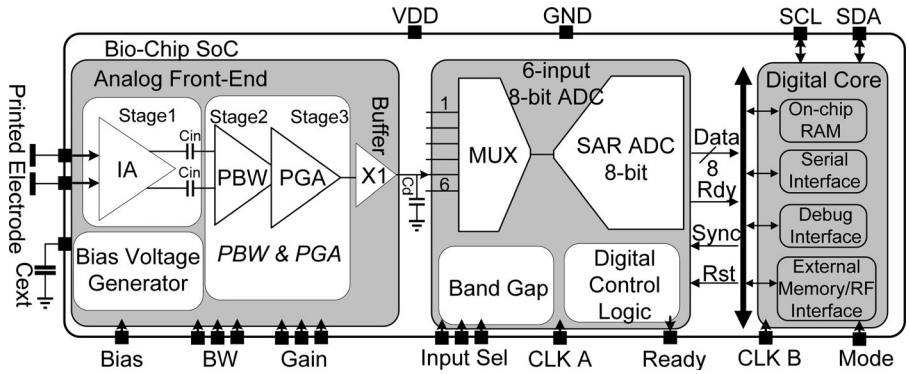

Fig. 2. Overall architecture of the bioelectric SoC.

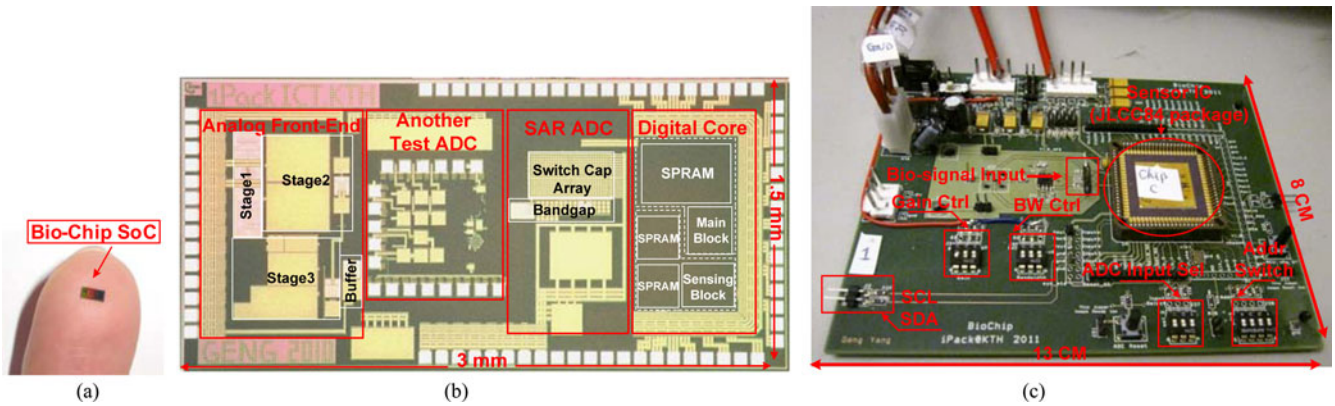

Fig. 3. (a) Fabricated bioelectric SoC on a finger. (b) Die microphotograph of the bioelectric SoC. (c) Measurement PCB.

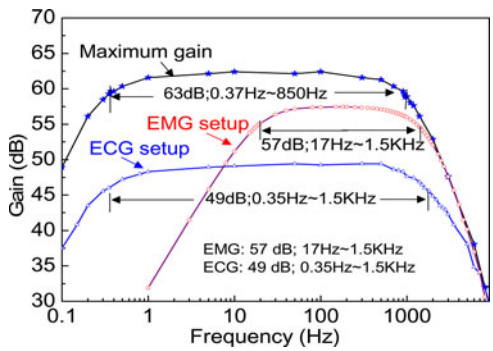

Fig. 4. Gain and bandwidth measurement of the analog front-end.

The high-pass frequency (HPF) is controlled by changing the value of the external capacitor $C_{\rm ext}$. To accommodate a specific biosignal, the SoC's 3-dB bandwidth can be configured between 0.3 Hz and 1.5 kHz by choosing a certain LPF and HPF. A unit gain buffer is implemented to drive a sampling SAR ADC in the following stage. The total capacitance of the capacitor array in the ADC is around 32 pF. Moreover, a common problem for a sampling ADC is the spikes generated by the switching of the capacitors during charging and discharging process, which can fold-over back into the front-end and correlate the output signal. Therefore, a decoupling capacitor C_d of 10 pF is applied at the buffer output, which effectively filters out the unwanted spikes.

By combining the three stages together, the front-end circuit can provide eight different close-loop gains from 45 to 63 dB. Fig. 4 illustrates the gain-bandwidth measurement results, where three gains are selected for demonstration. The gain of 49 dB with bandwidth between 0.35 Hz and 1.5 kHz is for ECG recording; and the gain of 57 dB with bandwidth from 17 Hz to 1.5 kHz is for EMG measurements. These gain-bandwidth setups are applied to the *in vivo* tests which are described in Section IV. The measurement results show that the analog front-end has a common mode rejection ratio (CMRR) around 90 dB, noise floor of 56 nV/$\sqrt{\rm Hz}$, and totally draws a current of 2.3 μA from a 1.2 V supply (including the unit buffer).

B. Six-Input 8-Bit SAR ADC

A complete biosignal acquisition SoC needs an ADC. In this design a six-input 8-bit charge-redistribution SAR ADC is implemented. It is mainly composed of five blocks: a bandgap, 6:1 MUX, a capacitive binary search array, a comparator and a digital control unit. The bandgap voltage generator provides a stable 800 mV reference signal for comparison. A key component of this SAR ADC is the capacitive binary search array that consists of 256 unit-capacitors (124 fF each). It acts as sampling capacitance in every sampling interval and as charge-sharing DAC during the conversion period. The common port (upper plate) of the capacitor array is connected to the noninverting terminal of the comparator. A symmetric identical dummy capacitor array is connected to the inverting terminal of the comparator to suppress the charge injection errors induced by CMOS switches during the comparison process. Three external pins control the

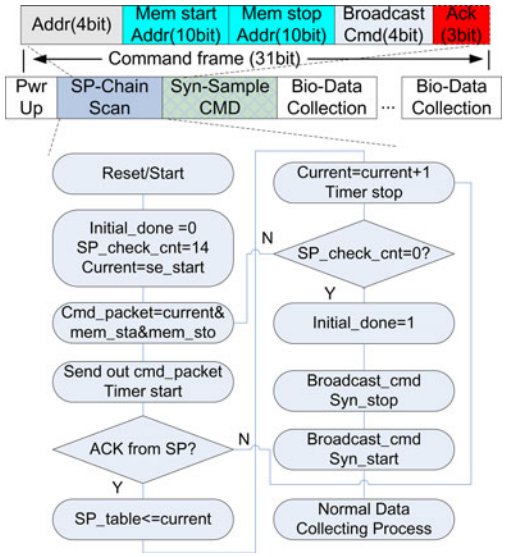

Fig. 5. Structure of a command frame and the algorithm of SPCS process.

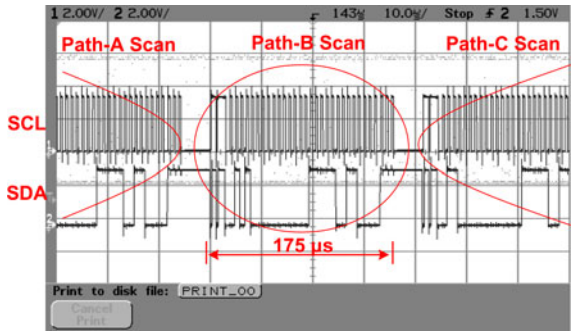

Fig. 6. Oscilloscope screenshot of the SCL and SDA during SPCS.

MUX to select one from the six input channels for A/D conversion. The input5 is reserved for the output signal from the analog front-end circuit; the other five inputs are available to interface other biology signals, such as body temperature etc.

The ADC provides a sample rate of 3 kS/s, working with an on-board oscillator of 100 kHz. The ADC consumes 1.8 μA from a 1.2-V supply.

C. Digital Core With Two-Wire Communication Protocol

Three pieces of single-port RAM are integrated on the chip that enables the Bio-Patch to buffer the digitized biosignal. Previous works on the digital core have been reported in [15] and [16]. The operating mode of the Bio-Patch can be configured via an external pin "Mode." If the pin is set to logic "1," the Bio-Patch is set as a MP; otherwise it is configured as a SP. The two operating modes differ in: a SP detects and processes the bioelectric signal, while for an MP, in addition to the bioelectric signal processing task, it also manages the whole network. Optional SPs can freely join in or move out from the network. The current version of the proposed e-healthcare system supports one MP and maximum 14 SPs (14 channels).

1) SP Mode: An SP processes the bioelectric signal with its analog front-end. It can activate/disable its ADC, and pick up the digitized biosignal from its A/D interface and store the data in its on-chip memory temporarily until they are periodically retrieved by the MP.

2) MP Mode: The MP initiates a command-based SP-chain scan (SPCS) process when the system is powered up or reset. Fig. 5 shows the structure of the command packet, and the control flow of the SPCS process. Fig. 6 shows the oscilloscope screenshot of SCL and SDA waveform during SPCS (175 μs for each SP scan). All SPs connected on the Active-Cable are scanned during this process. As a result, their address information is recorded in MP's on-chip buffer. After SPCS, the MP broadcasts a "Syn-Sample" command to activate the ADC located in each SP to startup synchronously. In this way, concurrent recording of multiple biosignals can be achieved. Since optional SPs may be added or removed according to the demands of a particular application, the network should be able to manage the case of a new SP joining in or an existing SP moving out. The FSM controller in the MP can cope with this issue by initiating the SPCS on a periodic basis of every 5 s to update the online SP information. If a new SP is detected, the "Syn-Sample" command will be resent for resynchronization. Data retrieving is performed every 0.03 s with a data rate of 350 kbits/s through the Active-Cable. Collected data are stored temporarily in the MP's internal memory. Finally, they are transmitted to the outside world via a wireless module. Bluetooth or custom designed wireless transmitter can be employed for wireless transmission. A novel all-digital polar transmitter is investigated for this design. It is implemented and tested on another chip, the measurement results are reported in [17].

The current consumptions of the digital core for the two operation modes are almost the same: 12.7 μA for the MP and 12.6 μA for an SP from a 1.2-V supply with 1-MHz system clock.

III. Printed Electrodes and Interconnections on Paper

Conventional Ag/AgCl wet electrodes are widely used in many bioelectric monitoring applications. However, signal quality degrades as the electrolyte gel dehydrates. Printed electrode fabricated using digital inkjet printing technology can be considered as a potential alternative to the wet electrode. The digital inkjet deposition is a contact-free manufacturing technology, and does not involve any complicated process, so passive elements can be easily made at relatively low cost [18]. Nanoparticle sliver ink NPS-JL (from Harima Chemicals) is directly printed on the paper substrate to form electrodes and circuit interconnections. A DMP inkjet printer (from Dimatix) is used to deposit small droplets (10 pL for each droplet) of silver nanoparticle ink directly and accurately on the paper substrate. The processing resolution is set to 1270 dpi with the drop spacing of 20 μm (center to center). The printed structure is sintered by heat in an oven so that the nanoparticles join together and form a continuous structure that enables formation of metal clusters/plate. In addition, sintering step can effectively remove the solvents and other additives present in the ink by

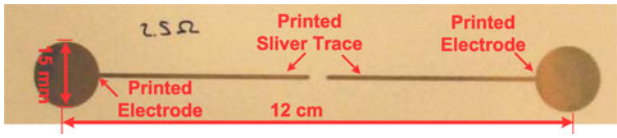

Fig. 7. Printed electrode using silver nanoparticle ink on a photo paper substrate.

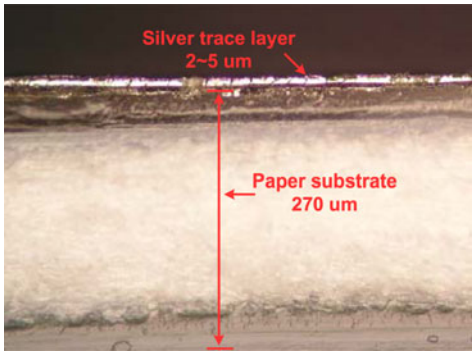

Fig. 8. Cross-sectional view of the printed electrode.

thermal decomposition and evaporation [19]. HP Advanced Photo Paper is selected as the substrate, since it is flexible, and features with smooth surface, great water, and smudge resistance as it is coated. It attaches well to the human body with respect to the skin convexity due to its flexibility.

Fig. 7 shows the inkjet-printed silver electrodes and connection traces on the photo paper substrate; the diameter of the electrode is 15 mm, and the distance between the two electrodes is 12 cm; the measured dc resistance of each connection trace is 2.5 Ω. Fig. 8 shows the cross-sectional view of the printed electrode. The thickness of the paper substrate is approximately 270 μm. The thickness of silver layer is approximately in the range of 2–5 μm. It is hard to clearly see the interface between silver and paper layers, due to the sintering and absorption of the ink to the paper. The conductivity of the printed trace is around 6 MS/m, which is 10% of silver bulk conductivity (63 MS/m). ECG and EMG signals are successfully captured by attaching the printed electrodes to the skin. The measurement details are given in Section IV.

In addition to the weak biosignal detected by the printed electrodes, digital signals can also be transmitted using printed silver traces. Previous work [20] has proved the digitized ECG signal can be successfully transmitted through the printed flat Active-Cable, where SCL and SDA are serial clock and data for serial communication.

IV. Implementation and *In Vivo* Tests

A. Performance of the Printed Electrodes

The performance of the printed electrodes is first evaluated by connecting them to the measurement PCB (the differential inputs of the SoC's instrumentation amplifier), as shown in Fig. 9. The gain-bandwidth setups illustrated in Fig. 4 are applied in the following ECG and EMG measurements. Fig. 9 shows the photograph of ECG *in vivo* test scenario using the printed electrodes and the standard measurement PCB. In this figure, the oscilloscope is stopped with captured ECG signal on screen, the printed electrodes are intentionally held outward of the chest (instead of toward the chest) for demonstration purpose. During monitoring, the electrodes are placed toward the chest, and an elastic bandage is used to apply mild pressure on the electrodes, so they are well attached to the skin. A clean ECG waveform (the output of SoC's analog front end) can be observed on the oscilloscope shown in Fig. 9. It should be noticed that no extra digital signal processing is applied for the removal of the 50 Hz noise. For EMG measurement, the same electrode pair is placed on the subject's biceps of right arm. The bandage is applied to fasten the electrode to the arm to minimize the motion artifact. EMG signal is detected when the subject contract and relax his biceps.

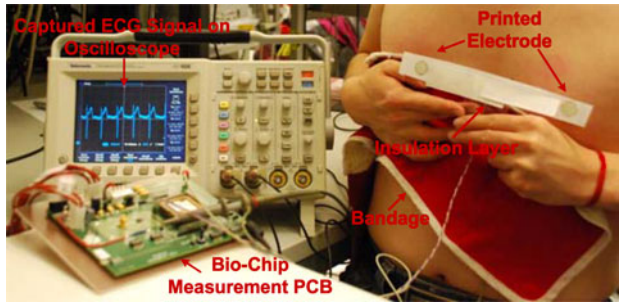

Fig. 9. Captured ECG signal using the printed electrodes and bioelectric SoC measurement PCB.

Fig. 10 presents the measured EMG and ECG signals and their fast Fourier transform (FFT) of the selected portion. The spectrum shows that most of the EMG energy locates in the frequency range from dc to 300 Hz; and the dominant energy is between 10 and 150 Hz, which is in line with [21]. For the ECG frequency spectrum, the dominant energy is below 20 Hz. Since the experiments are taken in a typical electrical engineering laboratory, a slight sign of 50-Hz power-line interference is observed in the FFT result.

The measurement results prove that the printed electrodes are adequate to detect the weak biopotential signal and can operate smoothly with the SoC sensor. After the performance evaluation of the printed electrodes, we move forward to a full integration solution by integrating the SoC sensor and the printed electrodes on the same substrate.

B. Bio-Patch Implementation

A prototype of the proposed Bio-Patch is implemented on a photo paper substrate, shown in Fig. 11, with a size of 6 cm $\times$ 8 cm. It is composed of three parts: a pair of printed electrodes, a soft battery, and the bioelectric SoC. Silver ink is directly printed on a photo paper substrate to make interconnections, thus a paper-based circuit board is formed. Previous work [22] has revealed the utilization of inkjet printing technology to successfully fabricate electrical interconnections between silicon chips and external components. A flat Enfucell battery is adopted in the Bio-Patch design. It is light in weight (1.4 g), and compact in size (60 $\times$ 42 $\times$ 0.7 mm^3). The soft battery is flexible and

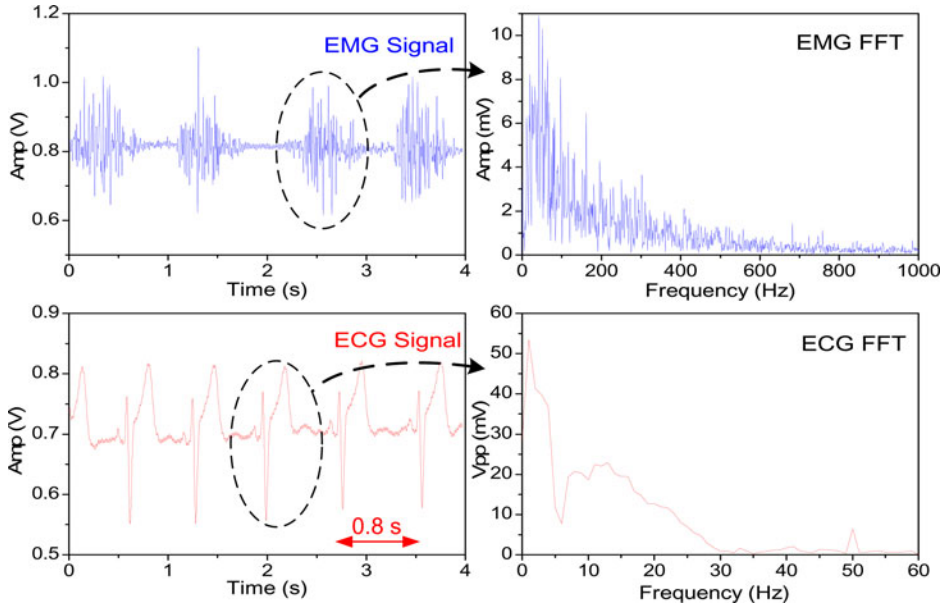

Fig. 10. Measured EMG and ECG signals using printed electrodes and the bioelectric SoC and their FFT analysis.

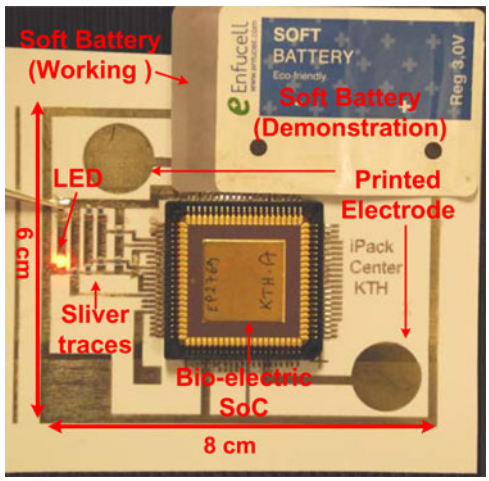

Fig. 11. Bio-Patch prototype on a photo paper substrate with printed electrodes, interconnections, bioelectric SoC and a soft battery.

TABLE I
PERFORMANCE SUMMARY

Parameter	*This work*
Silicon area	1.5 mm × 3 mm
Supply voltage	1.2 V ~ 3.3 V
Patch size	6 cm × 8 cm
Num. of patches supported	14
Ink	NPS-JL from Harima
Paper substrate	HP Advanced Photo Paper
Electrode	Printed silver electrode
Front-end power	2.3 μA@1.2 V
ADC power	1.8 μA@1.2 V
Digital core power	12.6 μA@1.2 V
Total power consumption	16.7 μA@1.2 V
Operation duration	Up to 2 months with a 3.0 V Enfucell soft battery

best suited for low power applications, such as microsensors, printing and new disposable applications. It provides a voltage source of 3.0 V with a battery capacity of 10 mAh. The Bio-Patch can operate up to 2 months with a single soft battery. As shown in Fig. 11, a red LED is turned ON after the soft battery is successfully connected to the Bio-Patch. It should be noticed that, in Fig. 11, two soft batteries stack on each other. The one facing downward is the actually working battery, and the one on top is intentionally placed face-up for demonstration. During experiments, a plastic cling wrap film is applied to tightly cover the patch as an electrical insulation layer to protect the circuit and components from skin-touch, while leaving two holes (with electrode size) above the printed electrodes to ensure the electrical contact with the body.

To achieve high level integration, on-chip circuit/devices are aggressively used to minimize the use of external components. Inks printed on the paper substrate and the materials of soft battery are the Restriction of Hazardous Substances Directive compliant. Therefore, Bio-Patch (except the bioelectric SoC) and its soft battery can be disposed of in normal household waste or recycled. The bioelectric SoC is plug-and-play; it is mounted onto the patch using a socket and it can be easily removed and reused in another patch. The performance of the implemented Bio-Patch is summarized in Table I.

Due to the loading effect, if the skin–electrode contact impedance is too large, the signal strength fed to input ports of the amplifier will be degraded [23]. This also leads to the degradation of signal to noise ratio, assuming the environmental noise is constant. Therefore, to minimize the loading effect, the bioelectric SoC is designed to have a high input impedance instrumentation amplifier. A PMOS input pair is employed to achieve large input impedance (4×10^{10} Ω @ 100 Hz). While the measured impedance of a wet electrode is around 15 kΩ @ 100 Hz, and 45 kΩ @ 100 Hz for the silver electrode [23].

Comparing with amplifier's large input impedance, the contact impedance of the wet electrode or printed electrode is insignificant. Hence, the loading effect can be well suppressed.

Fifty Hertz power-line radiation is a dominant source of electrical noise. The integrated instrumentation amplifier has a fully differential architecture and a high CMRR of 90 dB, which can effectively reject the common-mode noise. Therefore, only a small bump of 50 Hz interference shows up in the FFT result of the measured ECG signal. The captured raw ECG waveform using the printed electrodes shows a good signal quality, and it is adequate to detect many heart-related problems, especially arrhythmia. A shielding method employed in [24] offers a practical approach for power line noise shielding. It can also be used in the Bio-Patch design to reduce the penetration of electromagnetic fields. We realize the current package size for the SoC sensor, JLCC84 (3 cm $\times$ 3 cm), is the bottleneck for the further miniature of the Bio-Patch. This can be solved by adopting a package with less leads and smaller physical size in the future design, since currently many pins are for test purpose and can be removed in the next tape out.

V. Conclusion

A prototype of Bio-Patch for e-healthcare applications is implemented by integrating the fully integrated SoC sensor and the printed silver electrodes directly on a photo paper substrate using the inkjet printing technology. This prototype is for performance evaluation and concept demonstration purposes, where still several off chip components are employed. However, this is a step forward toward a completely fully integrated solution. The presented Bio-Patch is featured with low cost and long battery life. It can work alone and also support serial connection to other patches using the proposed flat Active-Cable for synchronous multichannel biosignal recordings. The command-based scan process realizes the real-time management of a scalable network. With the *in vivo* ECG and EMG recording experiments, we demonstrate the performance of the sensing and interconnection elements fabricated using inkjet printing technology, and evaluate the feasibility of integrating the SoC sensor with the printed electronics. The Bio-Patch-based hybrid biosensing system offers a promising solution for the future e-healthcare technology.

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