Paper II

G. Yang, J. Chen, L. Xie, J. Mao, T. Hannu, and L.-R. Zheng “A Hybrid Low Power Bio-Patch for Body Surface Potential Measurement” *IEEE Journal of Biomedical and Health Informatics (J-BHI)*, 2013 (accept).

Paper II

A Hybrid Low Power Bio-Patch for Body Surface Potential Measurement

Geng Yang, Jian Chen, Li Xie, Jia Mao, Hannu Tenhunen, and Li-Rong Zheng

Abstract—This paper presents a wearable Bio-Patch prototype for body surface potential measurement. It combines three key technologies, including mixed-signal System on Chip (SoC) technology, inkjet printing technology and anisotropic conductive adhesive (ACA) bonding technology. An integral part of the Bio-Patch is a low-power low-noise SoC. The SoC contains a tunable analog front end, a successive approximation register analog to digital converter, and a reconfigurable digital controller. The electrodes, interconnections and interposer are implemented by inkjet-printing the silver ink precisely on a flexible substrate. The reliability of printed traces is evaluated by static bending tests. ACA is used to attach the SoC to the printed structures and form the flexible hybrid system. The Bio-Patch prototype is light and thin with a physical size of $16 \times 16 \text{ cm}^2$. Measurement results show that low-noise concurrent electrocardiogram signals from 8 chest points have been successfully recorded using the implemented Bio-Patch.

Index Terms—body surface potential, Active Cable, inkjet printing, ACA, wearable device, SoC, Bio-Patch.

I. INTRODUCTION

WITH the developments in complementary metal-oxide semiconductor (CMOS) technology, fully integrated design has shown advantages in many advanced medical applications, in terms of low-power consumption and high-level integration. Examples can be found in battery-powered portable, wearable or even implantable medical devices [1] [2]. One of the current trends in healthcare systems is to move parts of healthcare services, such as routine check for senior citizens, from hospitals to home environments [3]. Thus release the limited hospital resources to the people in emergency needs. Home healthcare can not only lighten the workload of doctors and nurses but also allow people to self-monitor their health condition anytime. Thus patients can become actively involved

in their own care process [3]. On the other hand, the technology trends for wearable medical devices are to develop unobtrusive, miniaturized, integrated, networked, and long lasting medical sensor devices [3] [4].

Heart monitoring is an essential part in home healthcare, since heart disease is chronic process and is one of the most common causes for death all over the world [5]. Body surface potential measurement has proven to be a clinically relevant method that enhances diagnostic ability due to the larger number of sampling electrode positions on the thorax [6]. It can be obtained by processing the concurrently recorded multi-channel electrocardiogram (ECG) signals [6]. However, traditional ECG recording technology suffers from the massive connecting cables between the electrodes and the processing equipment. Examples can be found in a standard 12-lead ECG monitoring, where 10 recording electrodes are used. Additional electrodes may also be added to the standard set-up when increased spatial resolution is needed. This may cause to the following problems: 1) it is uncomfortable and inconvenient for the users if wires are worn around the body all day long; 2) the wires can easily get tangled; detangling is time consuming; 3) electrodes placement and configuration require professional assistance from nurses or healthcare professionals; any faulty connection may lead to a misdiagnosis. The facts above become limiting factors when considering the wider use of medical devices in home healthcare applications.

Based on these considerations, we develop a low power Bio-Patch for wearable body surface potential measurement. It is implemented by integrating a customized System on Chip (SoC) on a flexible polyimide (PI) substrate. The system is capable of concurrently detecting ECG signals from 8 different points across the heart. Conductive electrodes are assembled beneath the patch, which eliminate the connecting cables and avoid electrodes mis-configuration problem.

As a key part, a mixed-signal SoC with a full integration of processing modules is employed to fulfill the signal acquisition and processing. The advantages of using a SoC include significant power saving compared with its counterparts using off-the-shelf components. Also from physical dimension point of view, sensor size is a key factor which directly affects user's comfort. Some research groups make their efforts to develop sensor node using off-the-shelf components, while the resulting size is rather big [7, 8], since too many discrete components are employed, as a result, they are not suitable for wearable healthcare devices. Sensor device miniaturization can be realized by applying this compact SoC in the Bio-Patch implementation, since fewer external components are required.

Manuscript received June 15, 2012. This work was financially supported by Vinnova (The Swedish Governmental Agency for Innovation Systems) through the Vinn Excellence centers program.

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Fig. 1. Overall architecture of the proposed wearable Bio-Patch.

A serial communication protocol is implemented on chip. The output from each sensor node is a serial data stream which is transmitted from node to node along a 2-wire serial link. The integrated digital core enables the SoC to set up a network using a variety of unicast or broadcast commands. In addition, the noise resistance of the detecting circuit and power consumption of the whole system are the major considerations in the SoC design. To effectively reject the environmental noise, a sensing circuit with a common-mode rejection ratio (CMRR) of 90 dB is designed by employing a fully differential current balancing architecture. The total power consumption of the SoC is 20 μ W from a 1.2 V battery.

PI substrate is selected as the patch carrier, since it is low-cost, flexible (wearable), printable, thin and light [9]. High resolution inkjet printing is utilized to fabricate the printed devices [10]. Advantages of using this additive manufacture process include: less process steps, lower cost and better environmental compatibility, since inkjet printing technology can avoid the masks preparation and etching steps. Therefore, it is suitable for customized bio-sensing electrodes, lead frames as well as other patterns according to specific requirements [11]. A printed Bio-Patch is formed by assembling the inkjet printed interposer lead-frame, conductive electrodes, and printed circuit board on the PI substrate. Anisotropic conductive adhesive (ACA) is an environmentally friendly technique, which has been widely used in flip chip bonding applications with a wide range of substrate [12]. In this design, ACA is employed to bond the Bio-Electric SoC chip onto the PI interposer to form a sensor node. A soft battery from Enfucell is attached to the Bio-Patch as power supply.

The proposed hybrid wearable body surface potential acquisition system integrates silicon based electronics with printed electronics, and takes advantages of both [13]. The rest parts of the paper are organized as follows: system architecture is presented in Section II; SoC design and implementation are presented in Section III. Bio-Patch prototype assembly process and *in-vivo* tests are shown in Section IV. Finally, discussion and conclusion are made in Section V.

II. SYSTEM ARCHITECTURE

The proposed wearable Bio-Patch is shown in Fig. 1. It is attached on the subject's chest over the heart area, and can concurrently detect 8 points ECG signals. The patch is implemented on a flexible PI substrate, and composed of 9 sensor nodes. For each sensor node, a SoC is embedded inside. Due to the reconfigurable architecture of the digital core, each sensor node has two operation modes: the one located in the middle of the patch is configured as a Main-Node (MN), while the other 8 sensor nodes placed around are configured as Sensing-Nodes (SNs), each SN represents one sensing channel, as shown in Fig. 1. The MN manages the whole system, while the SN performs ECG signal acquisition. The detected ECG signal is stored locally in a SN's on-chip RAM before periodically collected by the MN.

A two-wire serial transmission protocol is implemented on-chip. It enables the serial communication between different sensor nodes. Therefore, a body area network is established over a group of sensor nodes. The output from each SN is serial data stream which is periodically retrieved by the MN and finally sent to a smart phone or a base station through MN's wireless interface. Commercially available wireless transmission module, such as Bluetooth, can be employed for wireless transmission. While a custom designed wireless transmitter, a novel all-digital polar transmitter, is also investigated for this design. It is implemented and tested on another chip, the measurement results are reported in [14].

An Active Cable [15] [16] is proposed to serially connect each sensor node to form a wired network. Its cross-section view is illustrated in Fig. 1. It is composed of five silver traces (PWR, GND, REF, SCL and SDA) with the thickness of 0.5 μ m by printing the silver ink on the PI substrate (thickness of 30 μ m). The Active Cable serves two purposes: 1) to share the power (PWR), ground (GND) and reference signal (REF); 2) to establish a wired network via a serial link (SCL and SDA). Comparing with the traditional star-topology full cable solution,

Fig. 2. Block diagram of the Bio-Electric SoC.

this serial connection method significantly reduces the number of connecting cables by sharing the signal chain among multiple sensor nodes, and hence improves users' comfort.

A typical ECG signal extraction flow can be described as follows: the MN provides its local electrical potential as a system reference (REF in the Active Cable). This REF signal is spread through out the whole patch to each SN via the Active Cable. While for a SN, REF is connected to one input of the AFE; a local voltage (detected from the electrode attached beneath the SN) is connected to the other input. The differential sensing is performed by subtracting the REF voltage from the time-varying local voltage. The extracted bio-signal is amplified and filtered by the front end circuit, and digitized by the on-chip analog to digital convertor (ADC). The digital core buffers the digital data temporarily in one on-chip RAM and transmits them out via the serial bus before the memory is full.

A common concern encountered by almost all the inkjet printed circuits on flexible substrate is the cracks in the printed lines caused by a large substrate bending. In order to ensure the system reliability, Active Cables employs a redundant architecture (Path A - Path D in Fig. 1) to connect the central MN with other SNs deployed around. Redundant connections allows reliable connectivity in cases where an electric connection is failed in some way. As long as 1 of the 4 redundant cables remains intact, the central controller can find its way to communicate with a specific sensing node, consequently, control the whole network. In order to further investigate the reliability of the printed Active Cable as well as other interconnections printed on the PI substrate, a bending test is performed in Section IV-B. The measurement results indicate that the printed silver traces can provide reliable electric connections, no crack happens even in the extreme case where the substrate is folded in half along the middle line.

III. BIO-ELECTRIC SOC

High level integration is accomplished by packing the essential signal processing blocks and control units on the same chip. The implemented Bio-Electric SoC contains three building blocks: 1) a tunable analog front end (AFE) circuit 2) a 6-input 8-bit successive approximation register analog to digital

converter (SAR-ADC) and 3) a reconfigurable digital core. The overall architecture of the SoC is illustrated in Fig. 2. The SoC is fabricated in a standard 0.18- μm Mixed-Mode 1P6M CMOS process and totally occupies 1.5 mm $\times$ 3 mm silicon area with a operating voltage ranging from 1.2 V to 3.3 V. The chip microphotograph is shown in Fig. 3.

A. AEF Readout Circuit and ADC

The ECG signal amplitude varies from one subject to another. In addition, variations of skin-electrode contact impedance, electrode size, and electrodes distance may also contribute to the fluctuation of the detected ECG amplitude. Therefore, to enable the detection of a wide range of bio-potential signals, a tunable gain is desired for the readout stage. Fig. 2 illustrates the block diagram of the programmable three-stage AFE. It allows us to adjust the gain of the amplification chain to an optimum value, while avoiding saturation distortion. Stage1 provides a fixed gain of 7. It is mainly composed of a current balanced instrument amplifier (CBIA) [17] and a common mode feed back circuit. The differential outputs of Stage1 are AC-coupled to Stage2. The close-loop gain of Stage2 is set to 12. Stage3 provides a tunable gain by tuning the equivalent capacitance of its inner capacitor bank via three external switches. The in-band gain of the Stage3 can be set to eight different values ranging from 2 to 19. The post-layout simulation results show that the three-stage front-end circuit can provide a tunable close-loop gain from 45 dB to 64 dB. The low-pass frequency of the front-end can be tuned to 80 Hz, 260 Hz, 400 Hz or 1 kHz by varying the load capacitance through the external switches. The high-pass frequency is set to 0.3 Hz. The AFE totally draws a current of 2.3 μA from a 1.2 V supply.

A SAR-ADC is employed due to its relatively good tradeoffs between power efficiency, conversion speed and resolution. A MUX is used to select one from six inputs to connect to the ADC. In this design, one input is reserved for the amplified bio-signal from the AFE; while the other five inputs are available for future sensor node's extension for other bio-signal sensing. The ADC provides a sample rate of 3.3 kS/s, working with an external clock of 100 kHz and consumes 1.8 μA from a 1.2 V supply.

Fig. 3. Chip microphotograph with building blocks marked.

B. AFE Measurement Results

The gain and bandwidth measurement results of the AFE are plotted in Fig. 4. The measured largest gain (63 dB) is slightly smaller than the simulation result (64 dB) which may due to the slight attenuation of the unit buffer located at the output port of the AFE. The gain or bandwidth can be independently adjusted via external pins. Totally eight different gain, ranging from 45 dB to 63 dB, can be selected (three of them chosen for illustration). For each selected gain, four user-programmable 3-dB bandwidths can be obtained using bandwidth switches. The upper part of Fig. 4 shows the case of 3-dB bandwidth adjustment with an almost consistent gain with the low cut-off frequency around 0.3 Hz and the high cut-off frequency adjusted from 85 Hz to 850 Hz. Regarding the SoC's electric characteristics, more details have been reported in previous work [18].

Due to its weak amplitude, the bio-signal is prone to be contaminated by the motion noise and other noise sources. In some applications, back-end digital signal processing (DSP) methods are applied to remove the out-of-band noise, however, these approaches consume extra power as well as silicon area. In this work, a fully differential current balancing architecture is applied in the AFE's design to achieve a CMRR of 90 dB, which enables the effective cancellation of the common-mode interference and also relieve the need of DSP. In order to measure input-referred noise (IRN), two input terminals are shorted to circuit ground and the output noise is measured with the signal analyzer (HP Signal Analyzer 35670A). The measured noise divided by the gain at its closet frequency point equals IRN. The IRN floor of -145 dB $V_{rms}/\sqrt{Hz}$ (56 nV/ $\sqrt{Hz}$) is observed.

C. Digital Core and its Operation

The on-chip integrated digital core includes three key features: 1) a reconfigurable control logic, which has 2 operation modes; 2) a 2-wire serial transmission protocol, which enables the establishment of a body area network; 3) and three pieces of RAM which allow the sensor node to buffer the digitized bio-signals. The performance of the fabricated digital core has been reported in [19] and [20]. An external pin 'Mode' is used to set the operation mode of each SoC. If the pin is set to logic '1', the node is configured as a Main-Node (MN); otherwise it is configured as a Sensing-Node (SN). In battery operated systems, great emphasis is put on power saving to extend the

Fig. 4. Gain-bandwidth measurement of the AFE.

battery life. Idle circuit blocks are switched off to minimize the power consumption. For example, the AFE and ADC of the MN is turned off for power saving. Further more, the digital core of the MN spends most of its time in a low-power sleep mode. For the two operating modes, the functionality of the MN and SN differs in:

SN mode: The main function of a SN is to sample, process and buffer the ECG signal. Each SN has a 4-bit address. The ECG signal from the subject's skin is amplified and digitized. Subsequently, the digital bio-signal is picked up and buffered temporarily in its on-chip RAM. If one RAM is full, the coming data are automatically switched to the second RAM. Before the on chip memory is full, the stored data will be periodically collected by the MN.

MN mode: The main role of the MN is to manage the whole network using a variety of commands. SN-Chain Scan (SNCS) process is initiated when the system is powered up or reset. During SNCS process, all SNs in the chain are scanned, the address of each active SN is stored in the MN's address buffer. Soon after SNCS is completed, bio-signal data collection process is initiated: the MN visits each SN in a serial manner. Collected data are buffered in a MN's on-chip RAM. Before the storage is full, the MN forwards these data to an external wireless transmission module, then it continues with the next round data collection.

The SNCS flow is described in Fig. 5a. The MN broadcasts a command packet containing a target SN address to the Active Cable. If the target SN is available, an acknowledgement frame is sent back immediately. Otherwise, if the target SN does not exist, no acknowledgement will be generated before timer overflow. Subsequently, the MN asserts the absence of the target SN, and continues with the next scan until all SNs are scanned. During this scan process, the ADC located inside in each SN stays in a 'Disabled' state. Fig. 5b illustrates the structure of command packet. Fig. 5c shows the screenshot of SCL and SDA on an oscilloscope during SNCS, where the scan of SN '1001' is enlarged for illustration. For the current version, the system supports one MN and up to 14 SNs. Once SNCS is completed, a synchronous sample ('Syn-Sample') broadcast command is initiated by the MN to trigger the SAR-ADC located in each SN to startup simultaneously. In this way, concurrent sampling of ECG signals from multiple chest

Fig. 5. (a) Command-based SN-Chain Scan (SNCS) process. (b) Packet structure for the command frame. (c) Oscilloscope screenshot of the SCL and SDA.

positions can be achieved. The average current consumptions of the digital core for the two operation modes are almost the same: 12.7 μA for the MN, and 12.6 μA for a SN from 1.2 V supply with a 1 MHz system clock.

IV. BIO-PATCH ASSEMBLY PROCESS AND IN-VIVO TEST

A. Assembly Process

Due to its high resolution and fine spacing, inkjet printing technology is utilized in this work to print conductive silver leads on PI substrate to form an interposer. Silver ink (from Sun Chemical Corp.) is directly printed by inkjet printer (DMP2800 from Dimatix). The volume of each drop is 10 pl and the drop spacing of inkjet processing is set to 20 μm (the pad pitch of the SoC is 90 μm). The printed leads exhibit high conductivity (typically around $1.25\text{E}+7$ S/m) and small thickness (around 0.5 μm for one layer printing). The electrical characteristics of the printed silver traces on PI substrate have been investigated and reported in [21]. Fig. 7a represents the printed PI interposer with a dimension of $18 \times 18 \times 0.03$ mm^3 . Cross-marks are made on the lower right corner for chip alignment during assembly. The integration of the tiny-size SoC on the printed interposer makes the sensor node miniaturization possible. The physical size of the interposer is much smaller and thinner than a standard JLCC-86 package ($30 \times 30 \times 3$ mm^3 , which was used as the chip container in previous tests). In addition, the JLCC package is not viable for this application due to its large and rigid nature, which makes the user uncomfortable if worn on human body.

ACAs have been widely used in flip chip packaging applications recently [22] [23]. It provides both electrical and

mechanical connections between chip pads and the substrate. The distinct advantages of using ACAs include reduced package size and thickness, improved environmental compatibility, and lower assembly temperature [24]. In addition, the process costs are lower compared with traditional bonding/soldering approaches, due to fewer processing steps [25]. Furthermore, ACA can be used in fine pitch applications as small as 40- μm pad width and 80- μm pad pitch. [26], [27]. Static and cyclic bending tests performed in [27] also indicate that the ACA joints are featured with sufficient reliability for consumer electronics.

In this work, the ACA from DELO is used to bond the SoC onto the interposer. Silver particles with a diameter of 1~3 μm are dispersed in the adhesive. A cross-section view of the overall flip chip assembly process is shown in Fig. 6. In the initial stage, the adhesive is not conductive in any direction, as conductive particles do not touch each other due to their tiny particle size and low density dispensed in adhesive, as shown in Fig. 6a. When pressure is applied, the conductive particles between the chip pad and substrate are deformed or breached, so the electrical contact can be made in vertical direction (perpendicular to the plane of the die), while in the horizontal direction it still keeps non-conducting, as shown in Fig. 6b. During the assembly process of this work, a thin layer of ACA is applied over the printed leads on the interposer (Fig. 7b). The flip chip is then aligned and attached onto the interposer using a fine placer LEICA-MZ75. During this packaging process, required pressure and heat are generated by the fine placer. The interposer with the attached SoC is shown in Fig. 7b. In order to ensure the electrical and mechanical contacts between the die and substrate, a protecting mold (medical compatible epoxy

Fig. 6. Assembly process using ACA. (a) Silicon chip alignment. (b) Thermo-compression bonding.

from EPOXY-Tech) is applied over the flip chip area, as shown in Fig. 7c. Finally, the interposer is flipped and attached to a larger PI mid-layer using ACA to form a sensor node, where several passive components are applied to configure the chip to a proper working condition, as shown in Fig. 7d. The sensor node is stand-alone with Active Cable interfaces printed on its four edges. It can be easily used for fast test or quick Bio-Patch assembly. Figure 8 shows the final Bio-Patch assembly which is accomplished by using conductive paste to mount the sensor node on the printed circuit board. The conductive electrode is formed by a thin silver plate on the PI substrate, where silver ink is inkjet printed on the PI substrate to form the electrode pattern and sintered in an oven at 145 °C for 1 hour. The fabricated silver electrode has a diameter of 15 mm, and it is attached

Fig. 7. SoC bonding and packaging. (a) Interposer on PI substrate. (b) Flip chip bonding with ACA. (c) Protecting mold. (d) Sensor node assembly.

Fig. 8. Bio-Patch final assembly on PI substrate.

beneath the Bio-Patch. The electric connection between the electrode and a sensor node (one input of the AFE) is made via through-hole on the substrate. The performance evaluation of the printed electrodes for ECG measurement (including contact impedance variation over frequencies, optimal electrode size and distance, and bending tests) has been reported in [28]. From the measurement results, it can be observed that the printed electrodes offer a comparable performance with the commercial pre-gelled electrodes.

The final assembled Bio-Patch prototype is shown in Fig. 9, with the size of 16 cm × 16 cm. The use of PI substrate and the integration of the bare die on the PI enable an ultra-thin patch solution with a total thickness less than 1.5 mm. 8 SNs are deployed and mounted on specific points of a printed flexible circuit board layer (corresponding to 8 positions on the subject's chest), and the MN is located in the middle of the patch. A flat soft battery from Enfucell (0.7 mm thickness, 1.4 g weight) is employed as the power supply. As shown in Fig. 9, the face-up battery is for demonstration purpose, the one facing downward is the actually working battery. Owing to its ultra-low power design, the power consumption is 15 μW for a MN, and 20 μW for a SN. The total power consumption of the proposed Bio-Patch (1 MN plus 8 SNs) is only 175 μW. In this case, the Bio-Patch (excluding the wireless module) has a battery life of around 7 days with a single 3.0 V 10 mAh Enfucell soft battery. Since wireless transmission module is recognized as a power-hungry component, it dominates the lifetime for the battery-operated wireless sensing devices. By using high-capacity lithium coin battery and low power wireless communication link (e.g. Bluetooth Low Energy), the proposed

Fig. 9. Bio-Patch assembly.

Fig. 10. Static bending test. (a) Test samples printed on PI. (b) Sharp folding.

Bio-Patch is capable of operating around 7 days for continuous monitoring. To protect the circuit, a thin plastic foil is applied to cover the patch. The Bio-Patch can be easily attached to a subject's chest by an elastic bandage or utilizing medical compatible adhesive at the bottom layer to ensure a good contact.

B. Reliability Test

Static bending test is performed to assess the reliability of the printed sliver traces on the PI substrate. 3 identical samples are printed on PI with a length of 4.5 cm and width of 2 mm, as shown in Fig. 10a. During tests, a multimeter is used to measure the DC resistance of the printed silver trace. We gradually decrease the bending radius of the printed samples to examine if metal crack or peel-off can occur under any condition. Throughout the entire experiment, no metal crack or peel-off occurs, the samples keep an almost consistent conductivity even when the radius is decreased to 4 mm (which means the sample is spiral-wound of 4 turns, a quite large bending comparing with the sample size). Finally, we make an extreme bending test by folding the sample hard to form a sharp edge, as shown in Fig 10b. From the figure, it can be observed that the connection of the silver trace keeps intact with the resistance of around 29 Ω . The measurement results prove that the printed Active Cable can offer a reliable interconnection even in the case of extreme folding.

A dynamic bending analysis of the inkjet-printed silver conductors on a polymer substrate is recently reported in [29], where the reliability and durability issues of the printed metal traces on flexible substrates are investigated. A 'bending machine' is applied in this dynamic bending test. The measurement results show that inkjet-printed traces prove to have good tolerance against bending, in addition, the durability and reliability of the inkjet printed silver traces are much better than their counterpart: etched copper samples [29].

C. Physiological Recordings

Initial measurement is made by applying the Bio-Patch prototype on the chest with the help of an elastic bandage. The concurrent multi-channel body surface potential recordings are conducted after the system completes the sensor node scan process SNCS. The synchronous sampling command ('Syn-Sample') is broadcasted to trigger the circuit embedded inside each SN to work simultaneously. This method ensures

Fig. 11. Obtained concurrent ECG signals. (a) ECG segment from SN-1. (b) Concurrent ECG from chest positions 1-8.

that the data sampled at each chest point are recorded at the same time instant. The front end read-out circuit located in each SN has the same configuration: with a gain of 60 dB and a 3-dB bandwidth of 0.3 Hz ~ 1 kHz. Fig. 11b shows the collected concurrent ECG data from SN positions 1-8, with a total time period of 12 seconds. It can be observed that the amplitude of the recorded ECG signal varies due to different sensing positions. To further verify the signal quality, a segment of ECG signal from SN-1 is enlarged for illustration in Fig. 11a. It can be observed that a clear ECG waveform is obtained. Since all the experiments are conducted in a typical electric laboratory, a slight 50-Hz power line noise can be observed.

For the long-term and homecare monitoring environment, the proposed Bio-Patch faces several challenges which may include: signal degradation resulting from the electrode damage due to the long-time skin-electrode contact; and the motion artifact which is generated from users' daily activities. Non-contact bioelectric signal measurement is an efficient approach [30]. With this method, the bioelectric signals (including the ECG signal) can be measured through the clothes/textile without direct skin-electrode contact, thus it can get rid of skin irritation and signal degradation problems. Also, the methods introduced in [31] and [32] can be applied in the Bio-Patch design to suppress the motion noise.

V. CONCLUSION

This paper presents a wearable hybrid Bio-Patch prototype utilizing System-on-Chip and high resolution inkjet printing technology for concurrent body surface potential measurement.

The integrated Bio-Electric SoC is featured with low power and low noise. It achieves 90 dB CMRR and 56 nV/ $\sqrt{\text{Hz}}$ INR density. The SoC fulfills the functional requirements of bio-potential sensing system, including signal sensing and processing, digitizing, data buffer and transmission. The tunable gain and programmable bandwidth enable the Bio-Patch to accommodate a wide range of bio-electric signal, and allow great freedom of choosing electrodes type, size and distance.

Instead of traditional standard package, an inkjet printed interposer on PI substrate is used as the silicon carrier. The distinct advantages of this approach include reduced package size, smaller thickness and more flexible to fit the body convexity. User comfort is improved by eliminating the cumbersome connecting cables with the sensor nodes serial connection method. The reliability of inkjet printed silver traces is evaluated through the bending test. The results indicate that the printed silver traces on PI substrate can be reliably employed in the fabrication of conductive electrodes, Active Cable and printed flexible circuit board. The system reliability is further enhanced by Active Cable's redundant architecture which can effectively reduce the risk of connection failure between sensor nodes. To evaluate the performance of the PI-base inkjet-printed Bio-Patch prototype, preliminary *in-vivo* test is applied. High quality concurrent ECG signals are successfully measured from 8 chest points. It is worth noting that up to 14 SNs are supported by the system. Additional sensor nodes may also be added when increased spatial resolution is needed. The proposed system is used for body surface potential measurement, yet it could also be adapted and applied for other wearable healthcare applications.

ACKNOWLEDGMENT

The authors would like to thank Dr. Fredrik Jonsson for his comments on Bio-Patch test.

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