A Wearable ECG Acquisition System With Compact Planar-Fashionable Circuit Board-Based Shirt

Jerald Yoo, *Student Member, IEEE*, Long Yan, *Student Member, IEEE*, Seulki Lee, *Student Member, IEEE*, Hyejung Kim, *Student Member, IEEE*, and Hoi-Jun Yoo, *Fellow, IEEE*

***Abstract*—A wearable electrocardiogram (ECG) acquisition system implemented with planar-fashionable circuit board (P-FCB)-based shirt is presented. The proposed system removes cumbersome wires from conventional Holter monitor system for convenience. Dry electrodes screen-printed directly on fabric enables long-term monitoring without skin irritation. The ECG monitoring shirt exploits a monitoring chip with a group of electrodes around the body, and both the electrodes and the interconnection are implemented using P-FCB to enhance wearability and to lower production cost. The characteristics of P-FCB electrode are shown, and the prototype hardware is implemented to successfully verify the proposed concept.**

***Index Terms*—Dry electrodes, electrocardiogram (ECG), electrodes, Holter monitor, planar-fashionable circuit board (P-FCB).**

I. Introduction

HEART disease is one of the major causes of death around the world. For example, in the United States, 26.6% of population died of heart disease, in the year 2005. An important aspect of the disease is that the sooner it is detected, the better quality of life the patient will have. Ambulatory electrocardiogram (ECG) monitoring is a good candidate for the detection of heart diseases; however, the difficulty comes from asymptomatic or intermittent properties of many chronic heart problems. Therefore, long-term monitoring is essential in detecting and treating the disease [1].

The first approach in detecting heart disease was the clinical ECG monitoring system [Fig. 1(a)], but this system is not appropriate for ambulatory monitoring because of the large size of the machine. Therefore, several studies attempted to continuously provide an ECG monitoring in everyday life. A good example of this is the Holter monitor system [Fig. 1(b)]. The subject wears and records his/her ECG data on 24-h basis. However, cumbersome wires make it inconvenient to use—there are typically six to ten wires worn around the body all day. Moreover, wet electrodes may result in skin irritation and signal degradation due to dehydration [2]. Thus, this approach is unsuitable for long term, continuous monitoring over several days. Meanwhile, a wearable health care system based on knitted integrated sensors was proposed [3]. It integrates sensors, electrodes and connections with conductive and piezoresistive yarn; it successfully proved the feasibility of fabric sensing elements and recording ECG signals with them. The only drawback of the system is that the production cost is high due to the knitting and interconnecting processes of the conductive yarn that have to be done additionally along with the conventional yarn, during the production of clothing itself. Digital plaster for the body area network (BAN) is another example [4]. A small patch integrates an ECG monitor with a battery and an industrial, scientific and medical (ISM) band wireless transceiver. This smart approach solves the convenience requirement of everyday life ECG monitoring: the subject can monitor the ECG data, and throw the sensor away afterwards. However, security and interference-resilient requirements are stringent for health monitoring, and, therefore, using an open-air wireless ISM band is unsuitable. Also, some people are reluctant to patch batteries directly on their skin all day.

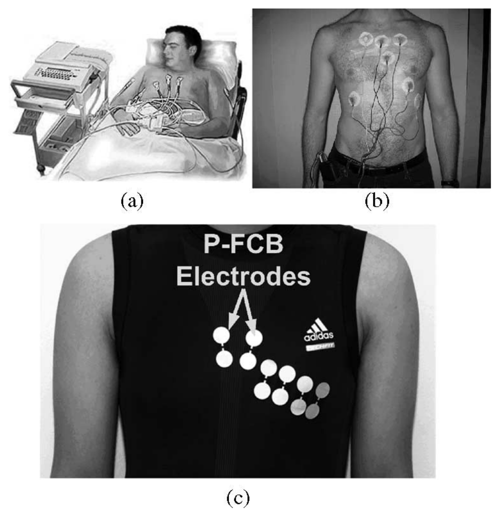

Fig. 1. (a) Clinical ECG monitoring system. (b) Conventional Holter monitor system. (c) Proposed ECG monitoring shirt with P-FCB electrodes.

Therefore, to solve the convenience and reliability problems, this paper presents a wearable ECG monitoring system with a compact planar-fashionable circuit board (P-FCB) technology-based shirt [5], [6] with low power consumption [Fig. 1(c)]. P-FCB technology enables dry electrodes and interconnection in the monitoring shirt at low cost.

This paper is organized as follows. Section II describes the system concept and the architecture of the proposed continuous ECG monitoring system. In Section III, P-FCB is examined

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J. Yoo, L. Yan, S. Lee, and H.-J. Yoo are with the Department of Electrical Engineering, Korea Advanced Institute of Science and Technology, Daejeon, Korea (e-mail: jerald@kaist.ac.kr).

H. Kim was with the Department of Electrical Engineering, Korea Advanced Institute of Science and Technology, Daejeon 305-701, Korea. She is now with the Interuniversity Micro Electronics Center, Leuven B-3001, Belgium.

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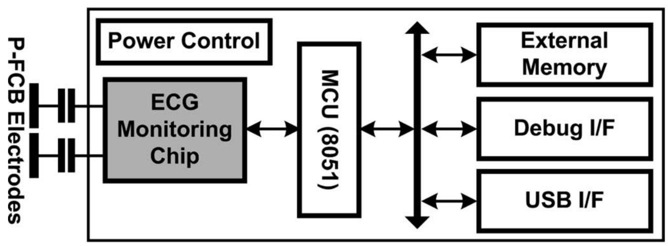


Fig. 2. Block diagram.

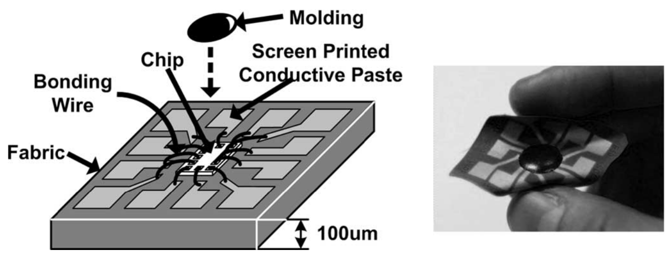


Fig. 3. Wire-bonded chip on fabric.

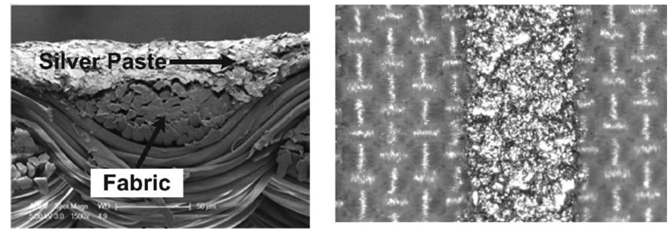
carefully, and the properties of dry P-FCB electrodes are described. Section IV shows in detail the system implementation and measurement results. Finally, Section V concludes the paper.

II. System Architecture

The proposed wearable continuous ECG monitoring system is composed of two parts: a group of P-FCB ECG electrodes and an ECG monitoring chip. For convenience, the monitoring shirt removes wires; interconnection is performed by P-FCB, and the electrodes are directly printed on the shirt itself, or the P-FCB electrodes on a small piece of a patch can be attached to the shirt as well. The basic concept is to integrate all the cumbersome wires and system into the clothes itself at low cost. The shirt-type clothes system is powered by a single 3.6 V, 1/2-AA size lithium battery, and the clothes can continuously operate and collect data from body up to 14 days without replacing the battery. To ensure continuous monitoring without stimulating skin, dry fabric electrodes are used at the sensors.

Fig. 1(c) shows the concept of the proposed ECG monitoring shirt. Six differential electrode pairs are placed around chest, as in standard chest lead electrode placement in conventional Holter monitor systems. In this figure, the P-FCB electrodes are intentionally attached outside of the monitor shirt (instead of inside) to show the concepts. During monitoring, the electrodes are placed on the inside, so they will be directly attached to the skin.

Fig. 2 is a block diagram of the proposed ECG monitoring system. Although this version of ECG monitoring system is implemented as a separate PCB board (except for the electrodes and the interconnections) because of debug interfaces and external memory, future versions will be directly integrated on the shirt as well as electrodes. Fig. 3 illustrates a chip directly wire-bonded on fabric. Then, a molding is applied, and as the molding hardens, its mechanical strength is sufficient to endure over 50 machine washings. More details on the P-FCB technology will be given in Section III.



(a) (b)

Fig. 4. (a) SEM image of the P-FCB cross section (1500× magnified). (b) Surface micrograph of the P-FCB (100× magnified).

The system is composed of P-FCB electrodes, the ECG monitoring chip, an 8051 MCU, an external memory, a debug interface, and a USB interface. The developed ECG monitoring chip includes an instrumentation amplifier with a programmable gain amplifier, an analog-to-digital converter (ADC), a compression accelerator, and an encryption block. The detailed architecture of the chip will be discussed in Section IV. Due to the P-FCB resolution (200 μm), the number of wire bonding pads is very limited, so the number of off-chip components must be minimized; therefore, on-chip devices are aggressively used. As a result, only nine off-chip capacitors and resistors are used to operate the ECG monitoring chip.

P-FCB allows the monitoring shirt to bend freely and to closely attach to the body, so a subject will feel comfortable during monitoring in everyday life. The ECG monitoring chip is connected to a differential electrode pair, and the ECG data will be captured and stored.

III. Dry Electrodes on Cloth

With chronic use, conventional Ag/AgCl wet electrodes show signal quality degradation as the electrolyte gel dehydrates. Moreover, there are toxicological concerns regarding electrolyte gels [2]. Therefore, to ensure continuous monitoring without harm, dry electrodes made out of fabric are used. In this section, P-FCB technology is shown, and dry P-FCB electrode characteristics are described in detail.

A. P-FCB

P-FCB is the planar printed circuit on fabric [5], [6]. Silver paste (Changsung Co. CSP-3163 or Fusikura Kasei Co. Dotite D-500) is directly screen printed on fabric to form patterned electrodes or a circuit board; the paste is a mixture of silver, polymer, solvent, polyester, and cyclohexanone to achieve high conductivity and adhesiveness [6]. Fig. 4(a) is an SEM photograph of the P-FCB cross section (1500× magnified), and Fig. 4(b) shows the surface micrograph of the P-FCB (100× magnified). The thickness of the paste is approximately 10–30 μm, and the minimum resolution of the process is currently 200 μm. The sheet resistance of the P-FCB ranges between 0.01 and 60 m$\Omega/\square$, as measured by a standard four-point probe.

Fig. 5 shows the P-FCB passive elements such as (a) an inductor, (b) a resister, and (c) a capacitor. The inductor has

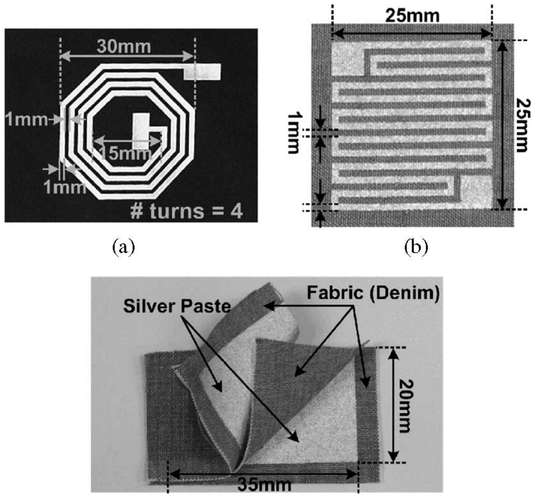


Fig. 5. Passive elements implemented by P-FCB. (a) Inductor (b) Resistor. (c) Capacitor.

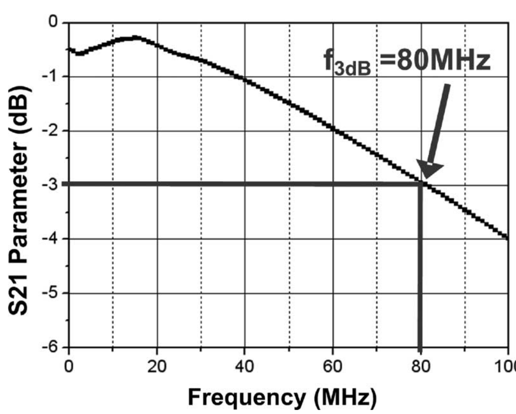


Fig. 6. Bandwidth of a P-FCB transmission line (15 cm long, 1 mm wide) [6].

an outer diameter of 30 mm, inner diameter of 15 mm, and the number of turns = 4. This inductor has $Q = 10.2$ and $L = 0.98\ \mu$H, and the wandering type resister has $R = 85\ \Omega$. Finally, the capacitor is composed of two 35×20 mm electrodes, and is measured to have $C = 10$ nF. These passive elements are essential in forming a complete system. P-FCB can also be used as a wire. Fig. 6 shows the bandwidth of a P-FCB transmission line 15 cm long and 1 mm wide, measured with a vector network analyzer [6]. Its bandwidth is 80 MHz, and this is sufficient to deal with bio signal processing.

Chips can be directly wire bonded and molded on fabric as well, as was shown in Fig. 3. Mechanical strength of the P-FCB is tested through washing tests; after 50+ times, electrical characteristics change is negligible [6]. As a result, the monitoring shirt can be easily made at low cost, and the wearability and flexibility are improved.

B. Dry P-FCB Electrode

An Ag/AgCl electrode with electrolyte is commonly used in ambulatory ECG monitoring [Fig. 7(a)]. The electrolyte gel forms a conductive path between the skin and an electrode, so

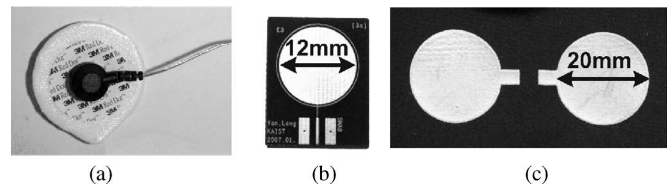


Fig. 7. Electrodes. (a) Wet-type (Ag/AgCl with electrolyte). (b) Dry type (metal contact). (c) Dry type (P-FCB).

the skin-electrode impedance is small, and stable reception of the signal is obtained. However, there are several reasons why the wet-type is inappropriate for ambulatory ECG monitoring over several days.

1) *Signal degradation.* As the electrolyte gel dries out, the skin-electrode impedance changes, and signal quality degrades. Re-applying the gel may mitigate degradation, but it is inconvenient to do so.
2) *Inconvenience.* Applying and removing the gel type electrode is unpleasant for both the patient and the clinician. This activity results in material and time costs.
3) *Safety.* There are toxicological concerns about the electrolyte gel within the Ag/AgCl electrodes; [7] indicates the gel ingredients may result in dermatitis.

To overcome these drawbacks of conventional wet electrodes for long-term monitoring, dry electrodes are considered as an alternative to the wet electrode [3], [8]–[11]. Fig. 7(b) shows the metal electrodes on the PCB board [8]. Attaching the electrode to the skin will successfully capture ECG data. This type of electrode is convenient and does not always cause dermatitis. However, since the electrode is coated on a stiff plastic board, it is uncomfortable to wear it over several days. A biopotential fiber sensor (BFS) successfully mitigates the stiffness by using coated fabric strands as an electrode [9]. Although the performance of BFS is good, its thin-film production process is complicated, including, for example, electrochemical deposition with the rapid dipping technique, and hence, too expensive. Moreover, it uses an inconvenient wired approach. A nonwoven fabric active electrode [11] is another good approach, but it still involves cumbersome wires, and it does not offer a direct chip integration solution.

P-FCB solves the above-mentioned problems simultaneously. Fig. 7(c) shows the proposed P-FCB electrodes. The silver paste is screen printed on fabric; the diameter of the electrode is 15–25 mm, and the distance between the two electrodes is 25–35 mm. Due to the flexibility of the fabric, it attaches well with respect to convexity of skin, and the effective contact area is increased, so the signal quality is better than that of [8]. This thick-film process does not involve any complicated process, so the production cost can be lowered compared to [9].

In dry electrodes, skin-electrode contact impedance is an important factor. If the impedance is too large, the loading effect degrades the signal strength that will be fed into the amplifier. Also, more noise will be induced. However, measurement reveals that the dry electrodes have approximately ten times higher contact impedance than the wet type. Fig. 8 shows the measured skin-electrode contact impedance of P-FCB and wet

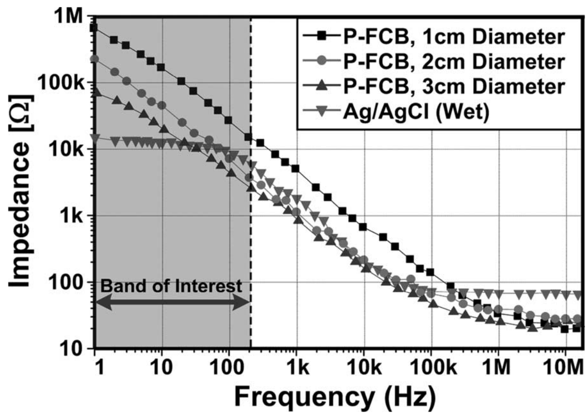


Fig. 8. Skin-electrode contact impedance of electrodes.

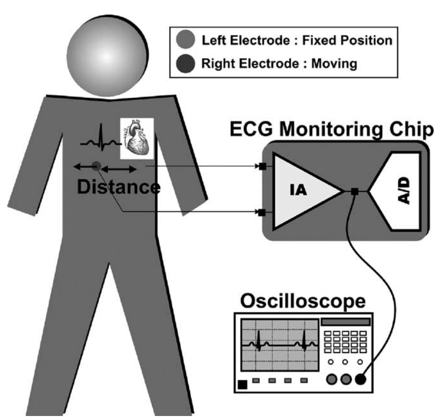


Fig. 9. Measurement setup for electrode distance effect.

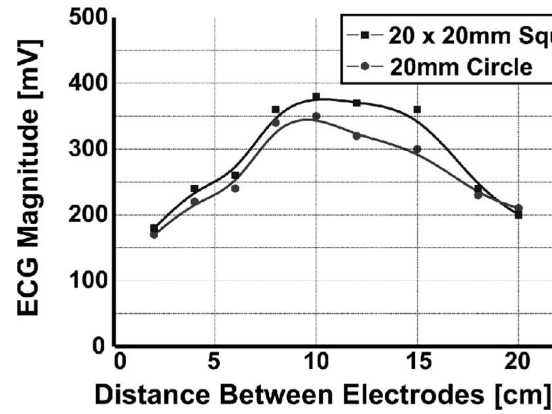


Fig. 10. Signal strength versus distance between electrodes.

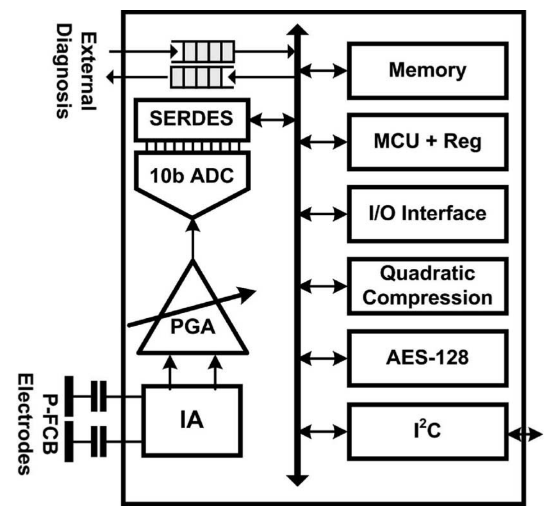


Fig. 11. ECG monitoring chip architecture.

electrodes. ECG signal bandwidth lies between 1 and 200 Hz, and the wet type has an impedance of around 2–20 kΩ within the band. On the other hand, contact impedance of a dry P-FCB electrode ranges between 10 and 800 kΩ, and the smaller the electrode size, the bigger the impedance. Therefore, to minimize the loading effect, the proposed system is designed to have higher input impedance at the instrumentation amplifier input than the conventional system.

The sensor patch has differential electrodes, as shown in Fig. 7(c). As the distance between the electrode changes, the captured signal strength varies. To find the optimal distance between the electrodes, the ECG magnitude after amplification (instrumentation amplifier output) is measured with respect to the distance between two electrodes. The measurement setup is shown in Fig. 9. Two P-FCB electrodes, both attached to the chest of a subject, are connected to a sensor patch, and the output of an instrumentation amplifier (IA) at the sensor IC is measured by an oscilloscope while altering the distance between electrodes. Gain of the IA is fixed to 200. Fig. 10 shows the result for a 20 mm diameter circle type electrode and a 20×20 mm square type electrode: the maximum signal strength is obtained at the distance of 10 cm, and if the distance is close enough, distance effect is negligible. Considering the chest size, the distance between the electrodes is set as 25–35 mm.

IV. Implementation and Measurement Results

Fig. 11 is the architecture of the ECG monitoring chip. It is composed of an instrumentation amplifier (IA) with a programmable gain amplifier, a 10 b SAR ADC, a compression accelerator, an AES-128 encryption accelerator, an internal memory, an MCU, *I/O* interfaces, and a pair of P-FCB electrodes. The electrodes are capacitively coupled to the IA input. Captured ECG signal is amplified at IA and fed into 10 b SAR ADC; the converted data is compressed with the quadratic level compression accelerator [5] and then encrypted, before being stored in the internal memory. For high security, AES-128 accelerator is used [12]. External memory can be used through the external memory interface if the internal memory becomes full. Internal memory is accessed by a fast-mode I^2C interface.

The chip is powered by a single 3.6 V, 1000 mAh, 1/2-AA size battery. When one channel is active, 3.0 mW minimum power consumption is measured, and this translates into 14 days of continuous ECG monitoring without replacing the battery. This is sufficient for 7-day ambulatory monitoring, which is essential for tracking heart disease [1].

The ECG monitoring shirt with the electrodes is implemented using P-FCB technology; electrodes and interconnections are

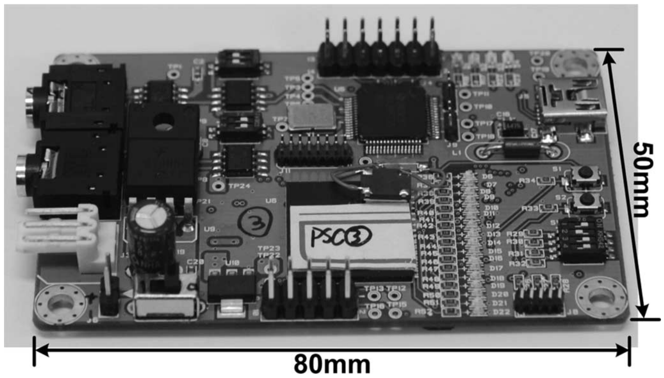
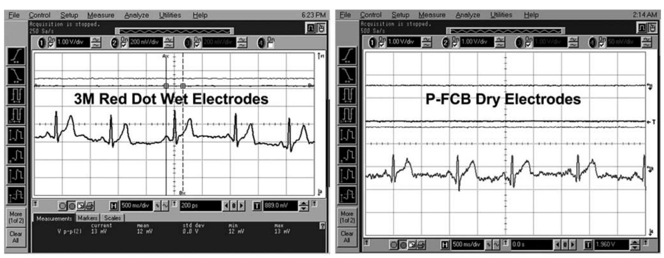


Fig. 12. Implemented ECG monitoring system board.



(a) (b)

Fig. 13. ECG signal captured by the system. (a) With wet electrodes. (b) With P-FCB electrodes.

provided by P-FCB, and the monitoring chip with subsidiary test blocks are implemented as a separate PCB board (Fig. 12). To minimize the motion artifact, mild pressure (below 20 mmHg) is required. Therefore, sportswear is selected so the shirt adheres closely to a subject's body, and, consequently, the electrodes will adhere to the body. The electrodes and the interconnection are printed on a nonstretchable cloth and are attached to the shirt via adhesives. When a subject wears the monitoring shirt, the electrodes are placed around the subject's chest, and ECG signal is captured with the monitoring chip. The proposed system is implemented as conventional PCB board, except for the electrodes and the interconnections between the board and the shirt.

Fig. 13 is the measurement result showing the monitoring system successfully capturing the ECG signal when the electrodes are attached to the chest. For comparison, both 3M Red Dot wet electrodes and the P-FCB dry electrodes are alternatively connected to monitoring chip. Fig. 13(a) is the signal captured with wet electrodes, and Fig. 13(b) is recorded with the dry electrodes (not at the same time, but at the same spot). The P-FCB electrode shows a little more distortion: since the P-FCB dry electrode has larger electrode impedance than that of a 3M Red Dot electrode, captured ECG signal is smaller in the P-FCB case, and therefore, the ECG signal is more prone to noise. Therefore, more gain at IA would mitigate the distortion. However, the captured ECG waveform with the P-FCB electrodes is sufficient to detect many heart-related problems, especially that of arrhythmia. Table I summarizes the performance and key features of the proposed system.

TABLE I
PERFORMANCE SUMMARY

Main Function	Continuous ECG Monitoring
Key Features	1) Wearable, Convenience 2) Dry Electrodes for Safety 3) P-FCB Low Cost Manufacturing
Supply	3.6V (½ AA Size)
Power Consumption	3.0mW
Operation Duration	Up to 14 Days of Continuous Operation with ½ AA Battery
Electrode	Differential P-FCB (Dry)

V. CONCLUSION

A wearable ECG acquisition system for ambulatory ECG monitoring over a very long period is implemented with P-FCB technology. The proposed system exploits a monitoring shirt with dry P-FCB electrode that removes cumbersome wires to increase wearability, and unlike conventional wet electrode, it is free from the skin irritation problem and, thus, is suitable for long-term monitoring. The prototype hardware was implemented to successfully verify the proposed system, and the measurement results were shown.

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Jerald Yoo (S'05) received the B.S. and M.S. degrees in electrical engineering from the Korea Advanced Institute of Science and Technology (KAIST), Daejeon, Korea, in 2002 and 2007, respectively. He is currently working toward the Ph.D. degree in the Department of Electrical Engineering, KAIST.

As a Chief Researcher at the Semiconductor System Laboratory, KAIST, he developed low-energy transceivers and wirelessly powered sensors for body sensor network, and an embedded processor for the parameter random access memory. His current research interests include low energy wearable body area network transceiver, wireless power transmission, and low-power biomedical microsystem.

Long Yan (S'07) received the B.S. and M.S. degrees in electrical engineering from the Korea Advanced Institute of Science and Technology (KAIST), Daejeon, Korea, in 2007 and 2009, respectively. He is currently working toward the Ph.D. degree in electrical engineering in the Department of Electrical Engineering, KAIST, where he has been engaged in the development of low power frequency shift keying transmitter for body channel communication, and low noise, wirelessly powered sensor for wearable body sensor network.

His current research interests include design of energy-efficient biomedical microsystem for body area network.

Seulki Lee (S'07) received the B.S. and M.S. degrees from the Korea Advanced Institute of Science and Technology (KAIST), Daejeon, Korea, in 2007 and 2009, respectively, both in electrical engineering. She is currently working toward the Ph.D. degree in electronic engineering in the Department of Electrical Engineering, KAIST.

Her research interests include the inductive coupling transceiver design and near-field communication for wearable computing applications.

Hyejung Kim (S'04) received the B.S., M.S., and Ph.D. degrees from the Korea Advanced Institute of Science and Technology (KAIST), Daejeon, Korea, in 2004, 2006, and 2009, respectively.

She is currently a Postdoctoral Researcher with the Interuniversity Micro Electronics Center, Leuven, Belgium. She developed low-energy biomedical signal processors, during her studies in KAIST. Her research interests include system-on-chip design for the continuous wearable electrocardiogram monitoring system and its implementations.

Hoi-Jun Yoo (M'95–SM'04–F'08) received the Graduate degree from the Electronic Department, Seoul National University, Seoul, Korea, in 1983, and the M.S. and Ph.D. degrees in electrical engineering from the Korea Advanced Institute of Science and Technology (KAIST), Daejeon, in 1985 and 1988, respectively.

He was engaged in the fabrication process for GaAs vertical optoelectronic integrated circuits during the Ph.D. studies. From 1988 to 1990, he was with Bell Communications Research, Red Bank, NJ, where he invented the 2-D phase-locked vertical cavity surface-emitting laser array, the front-surface-emitting laser, and the high-speed lateral heterojunction bipolar transistor. In 1991, he became a Manager of the Dynamic Random Access Memory (DRAM) Design Group, Hyundai Electronics, Ichon, and designed a family of fast-1M DRAMs to 256M synchronous DRAMs. In 1998, he joined the faculty of the Department of Electrical Engineering, KAIST, and currently is a Full-time Professor. From 2001 to 2005, he was also the Director of System Integration and IP Authoring Research Center (SIPAC), funded by Korean Government to promote worldwide IP authoring and its system-on-chip (SoC) application. From 2003 to 2005, he was the Full-time Advisor to Ministry of Korea Ministry of Information and Communication and National Project Manager for SoC and Computer. In 2007, he founded System Design Innovation and Application Research Center (SDIA), KAIST to research and to develop SoCs for intelligent robots, wearable computers and bio systems. His current interests include high-speed and low-power network on chips, 3-D graphics, body area networks, biomedical devices and circuits, and memory circuits and systems. He is the author of the following books: DRAM Design (Seoul, Korea: Hongleung, 1996; in Korean), High Performance DRAM (Seoul, Korea: Sigma, 1999; in Korean), Low-Power NoC for High-Performance SoC Design (Boca Raton, FL: CRC Press, 2008), and chapters of Networks on Chips (New York, NY: Morgan Kaufmann, 2006).

Dr. Yoo is an Executive Committee Member and the Far East Secretary for IEEE International Solid State Circuits Conference, and a Steering Committee Member of IEEE the Asian Solid-State Circuits Conference (A-SSCC). He was the Technical Program Committee Chair of A-SSCC 2008. He received the Electronic Industrial Association of Korea Award for his contribution to DRAM technology, in 1994, the Hynix Development Award, in 1995, the Design Award of Asia South Pacific Design Automation Conference, in 2001, the Korea Semiconductor Industry Association Award, in 2002, the KAIST Best Research Award, in 2007, the A-SSCC Outstanding Design Awards, in 2005, 2006, and 2007, and the Design Automation Conference/ISSCC Student Design Contests Student Design Contests Award, in 2007 and 2008.