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An ultra-low power (ULP) bandage-type ECG sensor for efficient cardiac disease management

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***Abstract*— This paper proposed an ultra-low power bandage-type ECG sensor (the size: 76x34x3 (mm^3) and the power consumption: 1 mW) which allows for a continuous and real-time monitoring of a user's ECG signals over 24h during daily activities.**

For its compact size and lower power consumption, we designed the analog front-end, the SRP (Samsung Reconfigurable Processor) based DSP of 30uW/MHz, and the ULP wireless RF of 1 nJ/bit. Also, to tackle motion artifacts (MA), a MA monitoring technique based on the HCP (Half-cell Potential) is proposed which resulted in the high correlation between the MA and the HCP, the correlation coefficient of 0.75±0.18.

To assess its feasibility and validity as a wearable health monitor, we performed the comparison of two ECG signals recorded form it and a conventional Holter device. As a result, the performance of the former is a little lower as compared with the latter, although showing no statistical significant difference (the quality of the signal: 94.3% vs 99.4%; the accuracy of arrhythmia detection: 93.7% vs 98.7%). With those results, it has been confirmed that it can be used as a wearable health monitor due to its comfortability, its long operation lifetime and the good quality of the measured ECG signal.

I. INTRODUCTION

To date, various vital signs sensors with the type of wearable smart shirts, a wristwatch, belts, and a patch have been proposed [1-4]. It is well known that those sensors must be small, lightweight, wireless, and capable of operating for long periods on a single battery charge (i.e. low power consumption) for user comfortability, while sustaining the accuracy [5]. Most of them, however, seem to be uncomfortable due to a tight-fitting to skin, relatively large size, and considerable weight.

With off-the-shelf components, it is very difficult to achieve key success factors of those sensors, such as small size, light weight, wireless operation, and low-power consumption etc. Therefore, we have developed the core components of those sensors, such as the AFE robust to various noises, the ULP DSP, and the ULP wireless RF communication chips. Also, we have designed the ULP bandage-type ECG sensor by adopting those core components.

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II. THE DESIGN OF THE BANDAGE-TYPE ECG SENSOR

As illustrated in Fig. 1, the bandage-type ECG sensor can be functionally divided into two sub-systems: the disposable electrode and the controller consisting of the AFE, the SRP (Samsung Reconfigurable Processor)-based DSP, the MCU (MSP-430, TI), and the ULP wireless RF (BAN).

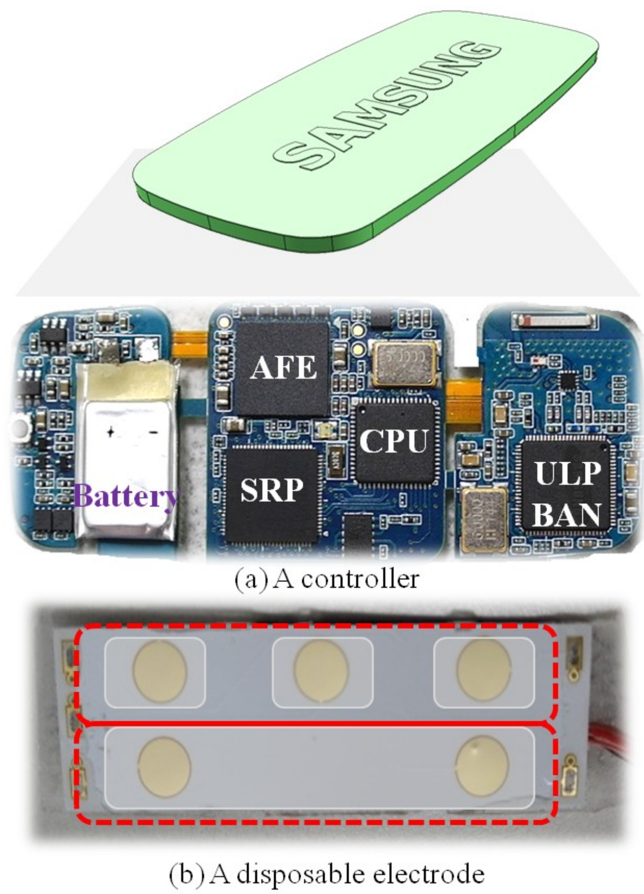


Figure 1 The bandage-type ECG sensor. (a) a controller and (b) a disposable electrode

A. *The AFE robust to noise*

To achieve the high SNR, the AFE has been designed to be robust to various noises, such as the flicker (1/f) noise of the CMOS transistor, the large DC electrode offset, the power interference noise, and motion artifacts. It has 3 identical readout channels whose channel consists of an ECG readout channel and a complex impedance readout channel. Each channel also has an AC current source of 2KHz frequency (Chopper clock generator) to incorporate a chopper modulation technique for the elimination of the flicker noise.

Also, it has been used to monitor the electrode-tissue impedance (ETI). The channel outputs are digitized by 5 SAR ADCs of which outputs are time-multiplexed on a master SPI output line.

In general, it is well known that the large differential DC electrode offset can be reduced by a high-pass filter (HPF) with the cutoff frequency of 0.5 Hz. In this paper, it is fully integrated into the AFE chip so as to avoid external components by the combination of the series capacitors and the bias resistors which are implemented with Metal-Insulator-Metal capacitors and switched-capacitor resistors, respectively. However, the HPF may limit the maximum achievable CMRR due to a mismatch of its inherent parasitic capacitance. To cope with this problem, thus, the HPF is complemented with a digitally controlled sub-fF accurate capacitor bank, consisting of 7-binary-scaled MOS capacitors to trim the mismatch of the parasitic capacitance.

B. The SRP-based ULP DSP

A local processing of the biomedical signals on a sensing node itself may be more preferable to the remote processing on a remote server in that it can minimize the amount of data communication, therefore reducing the radio power consumption. A DSP can play a critical role in detecting which event deserves to be analyzed by doctor. To detect an event from an ECG signal, first, R-peaks are recognized by the R-peak detection algorithm, then some features, such as a RR interval and a QRS width etc, are identified based on those R-peaks, and finally it is decided if there is any abnormal event on the basis of those features. To keep the overall power consumption of the sensor node very low, the DSP should allow low power operations as well. For the low-energy local signal processing, the SRP-based ULP DSP has been designed with TSMC 40nm process technology where the power and performance can be adapted according to application processing requirements.

The SRP-based ULP DSP is what is optimized for various biomedical applications as a variation of the ADRES processor [6]. It supports for two modes: the VLIW mode for more complex control flows and the CGRA (Coarse Grained Reconfigurable Array) mode for acceleration of a parallelizable loop based on software pipelining.

Biomedical applications usually spend most of their time in collecting data, in which data is transferred from the sensor to the memory by the DMAC (Direct Memory Access Controller). During this mode, the SRP-based ULP DSP is shut down. It starts data processing with the interrupt generated by the DMAC when the DCMC transfer is completed and/or the memory buffer is full.

Power Management Unit plays a major role to control all power domains (16 domains depending on the processing modes, the RMA etc.) as an effective way to save power by shutting off unused domains. As another means for energy optimization for specific applications, the SRP-based ULP DSP adopts the Dynamic Voltage Frequency Scaling by which the clock (from 1 MHz to 100 MHz) and the supply voltage (from 0.4V to 1.1V) can be set freely, so the power and the performance also can be tuned. That is, the SRP-based ULP DSP can be operated at 0.4V and 1MHz or at 1.1V and 100MHz.

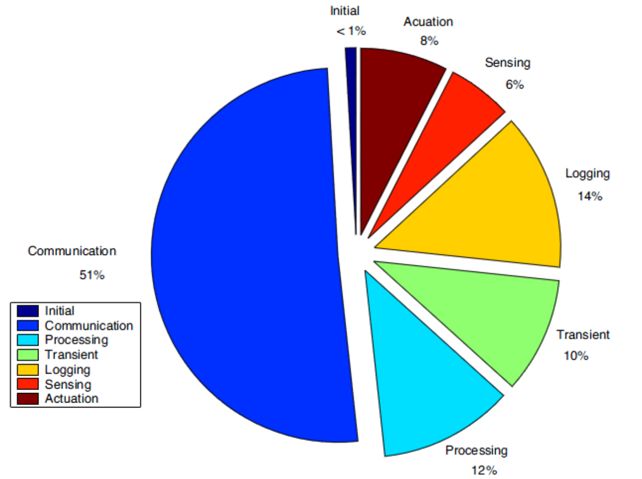


Figure 2 The power budget for a mobile vital signs sensing platform [7]

C. The ULP wireless RF (BAN)

As shown in Fig. 2, the RF component contributes to a large fraction (51%) of the system power consumption. The most commonly employed wireless communication technologies in BANs include the BlueTooth (IEEE 802.15.1) and the ZigBee (IEEE 802.15.4). However, in spite of their low power modes, they are not suitable for most mobile health monitoring because of large power consumption and the complexity of their protocol stack.

To resolve the power issue of the radio communication, we have taken two main approaches. One is to reduce the power consumption of the RF chip itself with the power efficient transceiver structure such as a super-regenerative receiver (SRO), a direct-modulated transmitter with on-off keying (OOK) and a pulse position modulation. The SRO makes the receiver architecture simple and power efficient by removing power hungry RF blocks, i.e., oscillator and mixer, in a heterodyne receiver.

With those approaches, it has achieved the high energy efficiency of 1 nJ/bit which shows lower power consumption as compared with the BlueTooth-LE (15nJ/bit @1Mbps and sensitivity of -83dBm) and the ZigBee (60~200nJ/bit @250Kbps and sensitivity of -94dBm), as shown in Fig. 3. It is operating in the medical BAN (2.36~2.4GHz).

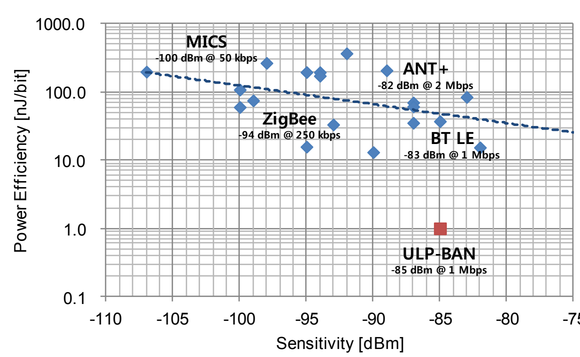


Figure 3 The power efficiency and sensitivity of various wireless Techs

The other is to decrease the RF transmission time by the local processing of the recorded signal run on the DSP, which detects any abnormal event and then sends to the RF module only the ECG segment of 4-min length, i.e. two 2-min segments before and after the detected abnormal event (called the event detection mode).

Figure 4 The typical ECG epochs for a patient with PVCs recorded using a Holter monitor and the proposed sensor

III. RESULTS AND DISCUSSION

The bandage-type ECG sensor as shown in Fig. 1 is a compact and power efficient sensor which has the size of 75x34x3 mm^3 and the power consumption of 1 mW due to the AFE, the ULP BAN chip and the SRP-based DSP chip.

The power consumption of the bandage-type ECG sensor is 1mW for the worst case where an event will occur every 4 minutes, while it is 500uW for the best case with only an event over 24 hours. This means that we can achieve great power reduction with the SRP-based DSP and the ULP BAN.

To evaluate validity of The Bandage-type ECG sensor in a clinical application, for 83 subjects we simultaneously recorded two ECG signals using it and a conventional Holter monitor, and then performed the comparison between them by the visual inspection of two experienced ECG technicians for an ECG epoch of 1 min length. Figure 4 illustrated the representative example of ECG epochs recorded with the conventional Holter monitor (on the upper trace) and the proposed sensor (on the bottom trace). In both of ECG tracings, the ECG expert can easily identify the existence of arrhythmias with the naked eye, which indicates that it can be sufficiently used as a continuous and real-time ECG monitor. As a result, its performance is a little lower as compared with the conventional Holter monitor, although showing no statistical significant difference (the quality of the signal: 94.3% vs 99.4%).

As mentioned above, we adopted the motion artifacts removal technique based on the HCP monitoring to improve the SNR of the ECG. Fig. 5 shows the ADF (Adaptive Filter) to estimate motion artifacts by using the HCP signal as its reference signal. The estimated motion artifacts are subtracted from the ECG signal contaminated with them, which results in a clean ECG signal.

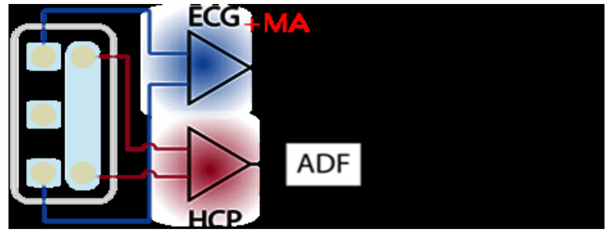


Figure 5 The adaptive filter to reduce MAs based on the HCP monitoring

Fig. 6 shows typical examples of the ECG signal (in red color) and the HCP (in green color). To assess its performance, we calculated the correlation between the MAs and the HCP, which showed the correlation coefficient of 0.75±0.18 (for 11 subjects), higher than that of the impedance monitoring technique. The adaptive filter of Fig. 5 was applied to an ECG signal from a subject during jogging and its results were shown in Fig. 7. This result indicates that the method can efficiently remove the MAs so that our R-peak detection algorithm accurately detected the R-peaks which were missed in the original ECG signal.

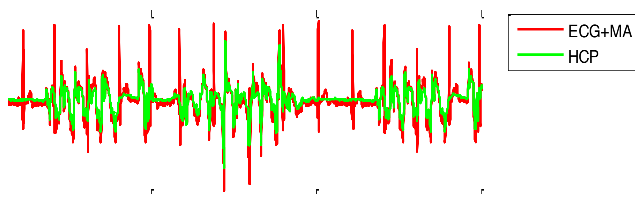
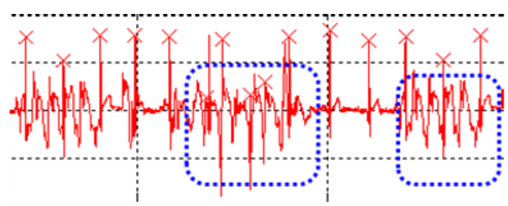
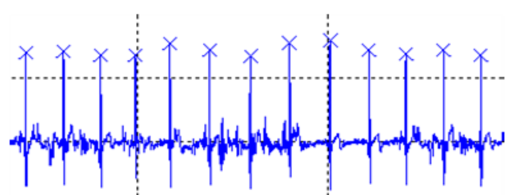


Figure 6 The ECG signal and the HCP



(a) The ECG signal contaminated with MAs



(b) The clean ECG signal with the ADF based on the HCP

Figure 7 The typical example of the adaptive filtering for MA suppression

IV. CONCLUSION

This paper presented the ultra-thinned bandage-type ECG sensor which allows for the continuous and on-line monitoring of a user's cardiac conditions. To remove some technical issues such as large size and limited energy autonomy of the current battery-powered devices, its core components has been developed which consists of the AFE robust to motion artifacts, the ULP wireless RF communication module, and the SRP-based ULP DSP.

Due to its small size, low weight, and low power consumption while remaining good quality of the measured signal, we believe that it is a good candidate for long-term, ubiquitous, on-line ECG monitoring. Therefore, it is expected that it can empower a user to cope with cardiovascular diseases by the early detection of abnormal cardiac events.

Its core components can be used not only for the remote patient monitoring but also for various wellness applications (e.g., a fitness and physical training, an obesity management, and a stress management etc.).

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