A $0.5\mu V_{\text{rms}}$ 12$\mu W$ Patch Type Fabric Sensor for Wearable Body Sensor Network

Long Yan, Jerald Yoo, Binhee Kim and Hoi-Jun Yoo

School of EECS, KAIST, 373-1, Guseong-dong, Yuseong-gu, Daejeon, 305-701, Republic of Korea
E-mail: yanlong@eeinfo.kaist.ac.kr

Abstract—A $0.5\mu V_{\text{rms}}$ 12$\mu W$ wirelessly powered patch type fabric sensor is presented for wearable body sensor network to continuously monitor personal bioelectric signals. Thick film electrodes are screen printed on the fabric with various metal components and their impedances of ~100k$\Omega$ are characterized. A 2-stage nested chopped analog readout front end (AFE) is optimized for the fabric sensor with reduced electrode referred noise performance of $0.5\mu V_{\text{rms}}$. A 10b folded SAR ADC reduces capacitive DAC (CDAC) size and relaxes the power budget of ADC driver by 94%. The proposed fabric sensor operates with system resolution of 9b and CMRR$>106$dB. The chip fabricated with 0.18$\mu m$ CMOS technology, the fabric sensor stacked by screen printed inductor (diameter=3cm and # turns=4) can measure the ECG and EMG signals with wirelessly transmitted power through inductive coupling.

I. INTRODUCTION

Battery-less wearable sensor opens up new applications both of the medical diagnosis and the personal continuous health monitoring [1]. The electrode for the wearable sensor and the data readout chip should be integrated on the same wearable platform. Previously, an electrolyte-less metal electrode was studied to maintain time-invariant contact impedance for a long term without allergic contact with the human skin [2]. However, the stiff interface with human skin makes it uncomfortable to be used in wearable application. A technology to integrate CMOS IC on a fabric directly was introduced for continuous healthcare [3]. With P-FCB, fabric electrode, inductor, and antenna can be fabricated and integrated with acquisition chip on the same fabric substrate for the wearable sensor. In this paper, the fabric electrodes are analyzed and an optimized bioelectric signal acquisition circuit will be presented. A thick film electrode pair is screen-printed with different metal components; silver powder, copper powder, and stainless steel powder. The electrodes with 1cm diameter and 2cm separation are screen-printed by P-FCB technology and their contact impedance is characterized in Fig.1. The metal powder forms a fine grained interface with skin to reduce motion artifact compared to the electrode introduced in [2] and resulting in stable contact impedance of maximum 100k$\Omega$ at $<200Hz$ in spite of its small contact area. However it contributes significant offset between 2 electrodes. It requires proper AC coupling before instrumentation to prevent the saturation of sensor readout circuits. In addition, the enhanced noise$<1\mu V_{\text{rms}}$, CMRR$>80$dB and PSRR$>40$dB performance of sensor readout circuits are also required to extract bioelectrical signal with resolution$>9$bit while operated in wirelessly powered environment.

II. ARCHITECTURE OF PROPOSED FABRIC SENSOR

Fig.2 shows the top architecture of the proposed fabric sensor chip and its detailed nested chopping front end. A self-configured wearable body sensor network (WBSN) controller introduced in [1] provides the required power wirelessly through inductive coupling channel ($Q=10.2$ and $L=0.98\mu H$) at 13.56MHz. Rectifier and low dropout regulator (LDO)
A low noise front end and its robustness to supply fluctuation are critical in wirelessly powered sensor design. In general, a single stage chopping is adopted at the sensor front end to suppress low frequency noise and to increase the CMRR/PSRR of the system [2,4,7]. However, it is challengeable to achieve required performance of noise and CMRR/PSRR by power constrained application. In addition, the residual offset caused by single high frequency chopping must be further suppressed to obtain required sensor resolution. In this design (Fig.3), an additional low frequency chopping is introduced to enclose the high frequency chopped amplifier [5] to realize a high SNR signal acquisition while operated in unstable power source. For the low electrode referred noise, NCA is realized by 2-stage amplification. In the first high gain stage, a diode connected load pairs give 22dB DC gain. A 10kHz demodulation clock is inserted in front of the second amplification stage (10dB) to realize 1/f noise suppression. A 10pF capacitor is added to the output of NCA for OTA-C filtering which reduces the 10kHz demodulation clock aliasing to the 625Hz chopping clock. At the input side, 625Hz filtering which reduces the 10kHz demodulation clock aliasing to the 625Hz chopping clock. At the input side, 625Hz modulation clock is pre-chopped with 10kHz to relax chopping spike by 1/16 than 10kHz single stage chopping. An enhanced power supply rejection of -48dB at 13.56MHz from LDO is realized by constant-\(g_m\) bias circuit with the MB5 (Fig.3) operated in the deep triode region. It behaves as a pseudo-resistor of 200k\(\Omega\) and its value is regulated by internal negative feedback loop to make the NCA independent of the supply variation.

**B. Programmable Gain and Bandwidth Amplifier (PGA)**

The proposed fabric sensor cooperated with WBSN controller [1] enables several fabric sensors to be put on at arbitrary locations of chest. In such application, different strength of bioelectric signals with different bandwidths can be monitored simultaneously which requires programmable gain and bandwidth stage. In previous design [2], a pseudo-differential implementation topology was presented to achieve variable gain of 6dB to 30dB and bandwidth of 300Hz–1kHz. However, it is not suitable in a low-power design because it needs two identical differential-single OTAs. In this design, a
fully differential OTA is implemented with switched capacitive (SC) common mode feedback (CMFB) as shown in Fig. 4. A non-overlapping clock frequency (P1, P2) of 80kHz is used to determine the output common mode of the PGA and it can be stabilized within 0.2ms. Mid-band gain control is realized by its ratio of the input capacitor and feedback capacitor from 16dB to 28dB and its high frequency cutoff can be tuned by the load capacitor to obtain a variable bandwidth of 100Hz~500Hz while consuming only 2μW.

C. Folded 10b SAR A/D Converter

Low energy consumption is essential in biomedical A/D converter design, especially in wireless powered application [1]. Among many types of A/D converters, successive approximation type utilized with CDAC is widely used due to its low power consumption. In order to realize 10b resolution, it conventionally requires total $2^{10}C_{\text{unit}}^2$ and driving such large size of capacitor usually requires a power consuming A/D driver. A reduced size CDAC topology [6] is a good candidate to meet small area and low power implementation. However, the $C_{\text{attenuation}}$ should be precisely determined to avoid code missing due to its limited linearity.

In this design, a folded CDAC topology (Fig. 5) is adopted to reduce capacitor size up to 94% compared to the design in [2] and the same size to the design in [6] while sacrifice only one additional conversion cycle. The CDAC composed of main CDAC and sub CDAC, and each consists of 6 binary weighted capacitor array to provide upper and lower 5bit A/D conversion code. In the analog input sampling phase, both the main and the sub CDACs sample the input value simultaneously. Upper 5bit is converted first and after 5 succeeding clock cycle, the approximated analog value is resampled by the sub CDAC to evaluate the lower 5bit with 1/16 scaled down reference voltage of upper 5bit evaluation. The total conversion rate is 12 clock cycle while dissipating only 2.2μW (10b resolution) which is 48% improvement compared with [2] (8b resolution). It is realized by 2 identical 5bit SAR control logic and 1 dynamic comparator.

IV. MEASUREMENT RESULTS

Fig.6 shows the input referred noise performance of the proposed AFE circuits. With the total differential gain of 60dB at the common mode input signal <300Hz, the CMRR of 106dB~120dB is obtained. The measured electrode referred noise power spectral density (PSD) shown in Fig.6 indicates that the noise floor is 47nV/√Hz which is equivalent to 0.5μVrms (0.5Hz~100Hz). It also satisfies the system requirements of >50dB signal acquisition. ECG and EMG are recorded with the proposed fabric sensor attached on the human chest with wirelessly powered operation as shown in Fig. 7. The measured peak SNR of ECG signal is 51dB and the maximum baseline fluctuation is <22% to ECG peak. The 10b folded SAR A/D converter gives an SFDR of 78dB and
Fig. 8. In Fig. 9, the linearity of 10b folded SAR A/D converter is shown, and the measured results shows that both INL and DNL within ±1 LSB.

The 1.3mm × 1.5mm sensor chip of Fig. 10 is fabricated with 0.18μm CMOS technology. Bare die is directly wire bonded on the fabric substrate and molded for protection. An electrode pair with 1cm diameter and 2cm separation is patterned besides the sensor chip by P-FCB technology introduced in [3]. In Table I, the performance comparison with previous works is summarized. It shows that the lowest noise level is achieved by NCA while dissipating only 12μW provided wirelessly.

TABLE I Performance Comparison

|--------------------|----------|----------|---------------------------|---------------------------|---------------------------
| Technology         | 0.18μm CMOS                        | 0.5μm CMOS             | 1.6μm CMOS                      | 0.8μm CMOS                      | 0.18μm CMOS                      |
| Power              | 22.5μA@1.8V                        | 20μA@23V               | 0.9μA@1V                        | 1μA@1.8–3.3V                      | 7.2μA@1.7V                      |
| Mid-band Gain      | 40dB (CMRR=100dB)                  | 20dB–68dB (CMRR=110dB) | 45.6dB–60dB (CMRR=140dB)        | 50.5dB (CMRR=100dB)              | 42dB (CMRR=100–120dB)           |
| ADC Performance    | ENOB = 7.6b INL≤±0.6LSB DNL≤±0.6LSB | —                      | ENOB = 10.2b INL≤±1.5LSB DNL≤±0.8LSB | —                                      | ENOB = 9.2b INL≤±1LSB DNL≤±1LSB |
| Input Referred Noise | 1.3μVRms @ (0.5Hz–200Hz)           | PSD = 57nV/√Hz 0.6μVRms @ (0.5Hz–100Hz) | PSD = 100nV/√Hz 0.98μVRms @ (0.05Hz–100Hz) | PSD = 47nV/√Hz 0.51μVRms @ (0.5Hz–100Hz) | —

V. CONCLUSIONS

A 0.5μVRms 12μW patch type fabric sensor is implemented with 0.18μm CMOS technology for wearable body sensor network. It is optimized for the fabric electrode in terms of the contact impedance to the skin with various metal components and battery-less operated environment. A nested chopped front end is realized for the low noise strategy and a 10b folded SAR A/D converter is designed for the low power consumption. Compared with the other state-of-the-art design, it can maintain the noise performance of 0.5μVRms while consuming only 12μW provided by wirelessly through fabric inductive coupling at 13.56MHz. With the help of the proposed patch type fabric sensor, continuously monitoring the personal health condition is possible with a peak SNR of 51dB at arbitrary location on the body without numerous batteries.

REFERENCES