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# A 0.6V 3.8 $\mu$ W ECG/Bio-Impedance Monitoring IC for Disposable Health Patch in 40nm CMOS

Jiawei Xu<sup>1</sup>, Qiuyang Lin<sup>2,3</sup>, Ming Ding<sup>1</sup>, Yao Li<sup>2</sup>, Wouter Serdijn<sup>2</sup>, Chris Van Hoof<sup>3</sup>, Nick Van Helleputte<sup>3</sup>

<sup>1</sup>imec - Holst Centre, Eindhoven, The Netherlands,

<sup>2</sup>Delft University of Technology, Delft, The Netherlands, <sup>3</sup>imec, Leuven, Belgium

**Abstract**—Simultaneous measurement of Electrocardiogram (ECG) and bio-impedance (BioZ) via disposable health patches is desired for patients suffering from chronic cardiovascular and respiratory diseases. However, a sensing IC must consume ultra low power under a sub-volt supply to comply with miniaturized and disposable batteries. This work presents a 0.6V analog front-end (AFE) IC consisting an instrumentation amplifier (IA), a current generator (CG) and a SAR ADC. The AFE can measure ECG and BioZ simultaneously with the single IA by employing a orthogonal chopping scheme. To ensure the IA can tolerate up to 300mV<sub>pp</sub> DC electrode offset and 400mV<sub>pp</sub> common-mode (CM) interference, a DC-servo loop (DSL) combined with a common-mode feedforward (CMFF) loop is employed. A buffer-assisted scheme boosts the IA’s input impedance by 7x to 140M $\Omega$  at 10Hz. To improve the BioZ sensitivity, the CG utilizes dynamic element matching to reduce the  $1/f$  noise of the output current, leading to 35m $\Omega$ /√Hz BioZ sensitivity down to 1Hz. The ADC shows a 9.7b ENOB when sampled at 20ksps. The total power consumption of the AFE is 3.8 $\mu$ W.

**Keywords**—ECG; bio-impedance; instrumentation amplifier; low noise current source; low supply

## I. INTRODUCTION

Chronic cardiovascular-respiratory diseases, like congestive heart failure (CHF) and obstructive sleep apnea (OSA), require long-term, continuous and comfortable monitoring of ECG and bio-impedance (BioZ) to detect abnormal heart rate, respiration and body fluid volume. For miniaturized, lightweight and low-cost disposable patches, alternative power sources such as organic paper batteries, 3D printed batteries or thermal energy harvesters are more interesting than bulky Lithium-ion cells. However, these promising batteries usually have a low output voltage, which would require circuits also operating at sub-volt supplies to avoid excessive power management losses of boost converters. Furthermore, a low supply AFE enables better co-integration with digital cells to facilitate power-efficient and on-the-node signal processing.

State-of-the-art IC solutions do not meet these requirements at the same time. Ultra-low power ICs [1]-[4] operating at 0.5-0.6V do not support BioZ and they are compromising on noise performance (i.e. [2][3] don’t meet the noise requirement of <30 $\mu$ V<sub>pp</sub> defined in ANSI/AAMI/IEC60601-2-47), while high-performance multimodal ECG/BioZ ICs [5][6] typically have 1.2-1.8V supplies and consume more power (>50 $\mu$ W/channel).

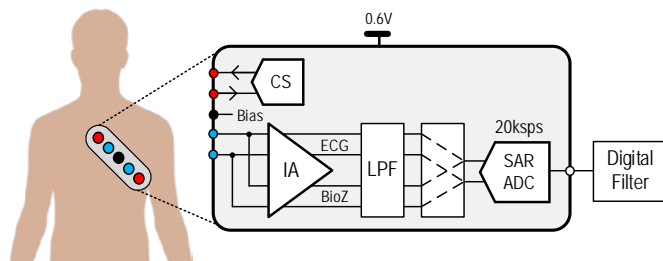


Fig. 1. IC block diagram and simultaneous ECG and BioZ measurement with one single amplifier

This work presents a 0.6V 3.8 $\mu$ W AFE (Fig. 1) including an instrumentation amplifier (IA), a BioZ current source (CS) and a SAR ADC to facilitate simultaneous monitoring of ECG and BioZ. An orthogonal frequency modulation scheme [7] enables power-efficient ECG and BioZ measurements with a single IA. To cope with large electrode-offsets and common-mode mains interference on the 0.6V low supply, a DC-servo loop (DSL) combined with a common-mode feedforward (CMFF) path is proposed. This allows the IA to tolerate up to 300mV DC electrode offset (DEO) and 400mV<sub>pp</sub> input CM fluctuation, respectively. BioZ measurement is enabled by a wide-swing and low-noise current generator equipped with regulated current mirrors and dynamic element matching (DEM).

## II. CIRCUIT IMPLEMENTATION

### A. Orthogonal Frequency Modulation

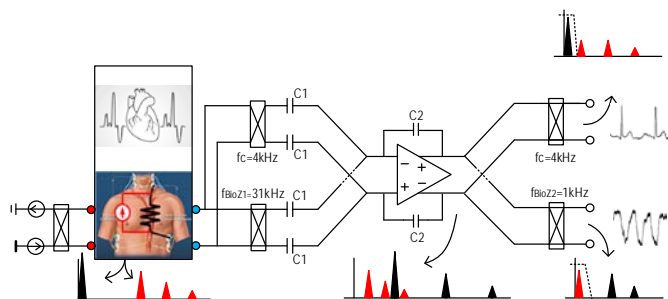


Fig. 2. Simultaneous ECG and BioZ measurement with one single amplifier

While traditional multimodal IC implementations rely on dedicated ECG and BioZ readouts, this work proposes a single amplifier-based ECG and BioZ readout (Fig. 2). A capacitively coupled IA (CCIA) concurrently measure both signals from the same electrodes by making use of signal properties: the BioZ is

measured by injecting an AC current above 1kHz. Hence, the ECG and modulated BioZ signals appear in different frequency bands before entering the IA. However, medically relevant BioZ spans from 1kHz-128kHz, this would impose strict BW requirements and high power on the readout [6]. To overcome this issue, the modulated BioZ signal is firstly down-converted to 1kHz (Fig. 2) before the CCIA and then demodulated further to DC at the output of the CCIA. The ECG signal is chopped at 4kHz, such that the fundamental and harmonic components of ECG and BioZ signals during amplification are located at different frequency bins in an orthogonal manner. At channel outputs, both ECG and BioZ signals are modulated back to the baseband respectively without interfering each other. The SAR ADC is oversampled at 20ksps to avoid folding of noise and residual harmonics. This also relaxes the design of anti-aliasing LPF in terms of its bandwidth and order. Sharing one CCIA for both ECG and BioZ channels improves power efficiency and reduces chip area, while the orthogonal frequency modulation ensures more than 60dB signal isolation between channels.

### B. Instrumentation Amplifier (CCIA)

The biggest design challenge for a 0.6V bio-amplifier is to ensure almost rail-to-rail input and output dynamic range in the presence of large external signals (300mV DEO, baseline drift and mains CM variations). This design utilizes the CCIA (Fig. 3) because it requires near-zero voltage swing at the virtual ground. To improve power-efficiency of the CCIA, the core amplifier is based on an inverter-based input stage [3] (Fig. 4) and a class A/AB output stage with switched-capacitor CMFB. Since chopping at the virtual ground node of a CCIA increases the noise, the choppers are implemented around the capacitive feedback network. However, this modulates the DC signals at the same time, and coupling capacitors  $C_1$  would fail to reject the DEO. To solve this issue, a DSL is provided to compensate the DEO (Fig. 3). The DC voltage at ECG output is tracked by a Gm-C integrator and a compensation current at  $f_c=4\text{kHz}$  is fed back to the virtual ground via  $C_{fb}$  to null the DEO current. The Gm has a complementary input (Fig. 4) to support CCIA's rail-to-rail output swing, and is chopped to reduce residual 1/f noise.

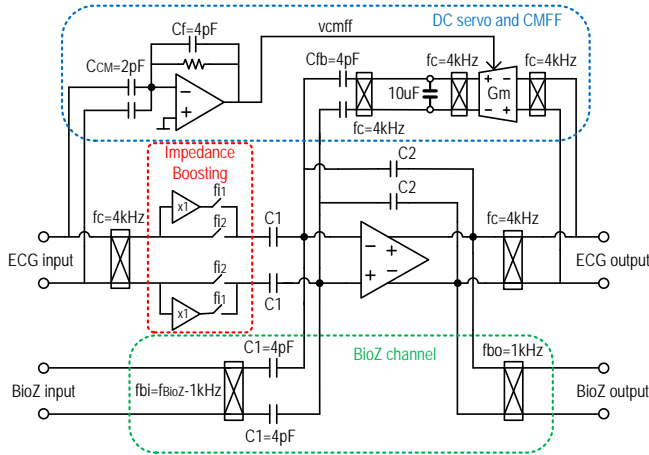


Fig. 3. Block diagram of the 0.6V ECG/BioZ instrumentation amplifier

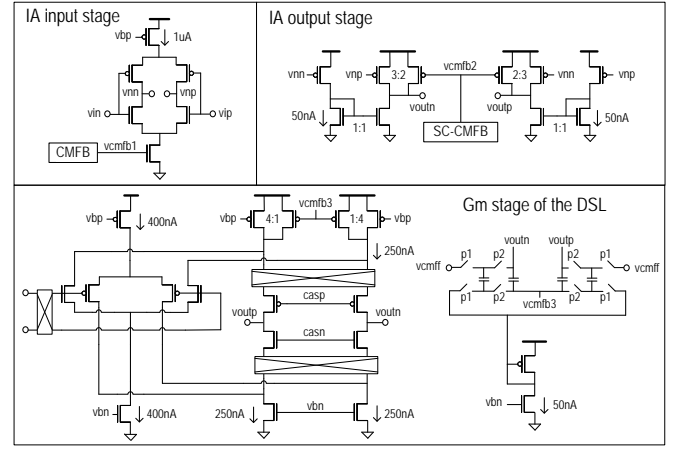
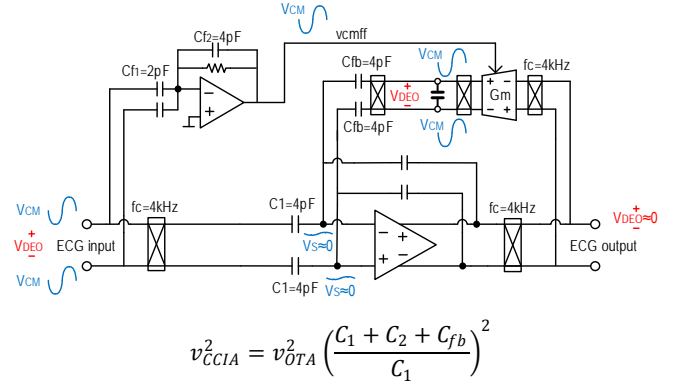


Fig. 4. Schematic of IA including inverter-based input, class A/AB output, Gm stage for DSL and CMFF

Although inverter-based input stages are attractive for low-power, they suffer from distortion with the presence of a large CM input signal. This is problematic for wearable biomedical applications where CM interference can be significant. Hence, a CMFF loop like [8] is used to reduce the CM swing at the virtual ground for improved linearity (Fig. 5). However, in this work, the input CM is fed forward to the virtual ground via the DSL's SC-CMFB reference (i.e.,  $v_{cmff}$  in Fig. 4) and  $C_{fb}$ , instead of adding another feedback loop [8]. Hence, the noise due to multiple feedback paths is reduced.



$$v_{CCIA}^2 = v_{OTA}^2 \left( \frac{C_1 + C_2 + C_{fb}}{C_1} \right)^2$$

Fig. 5. CMFF combined with DC servo for noise reduction

The chopping CCIA's suffer from limited input impedance formed by SC resistors. Prior work employed positive feedback loops to boost input impedance [3][9] but they may suffer from instability and the practical boosting factor heavily depends on parasitic capacitance. In this work, the input impedance of the CCIA is boosted by two pre-charging buffers placed after the input chopper (Fig. 6), this is similar to [8] but with a different clocking scheme. The buffers are periodically connected to the signal path for  $15.625\mu\text{s}$  whenever the chopping clock switches. Hence, the spike current to charge  $C_1$  is provided by the buffers, instead of ECG source input. This reduces the net current draw from the source and thus improves the input impedance over the entire bandwidth. In addition, this approach eliminates the instability risk. Thanks to the duty-cycling buffers, their noise contribution is negligible.

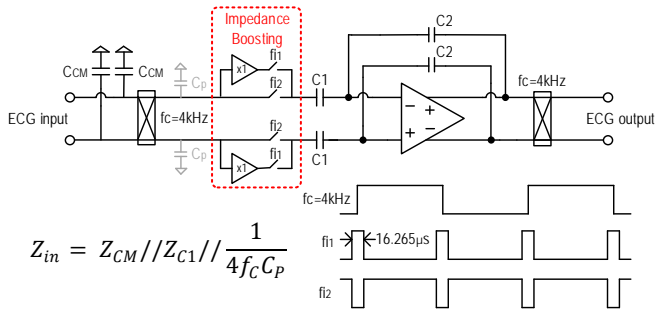


Fig. 6. Input impedance boosting with auxiliary buffers

### C. BioZ Current Generator

The BioZ current source (CS) is shown in Fig. 7. It has an output magnitude of  $5\mu\text{A}$ - $100\mu\text{A}$  and an output frequency of  $4\text{k}$ - $128\text{kHz}$ . Since most of BioZ activities (e.g., respiration, body fluid volume) are below  $10\text{Hz}$ , a main design challenge is to reduce the CG's  $1/f$  noise for improved sensitivity. Apart from noise, achieving a large compliance range under  $0.6\text{V}$  supply is also important. This ensures that the CS remains operational when considering voltage drop over the electrode impedance, which is typically larger than the BioZ.

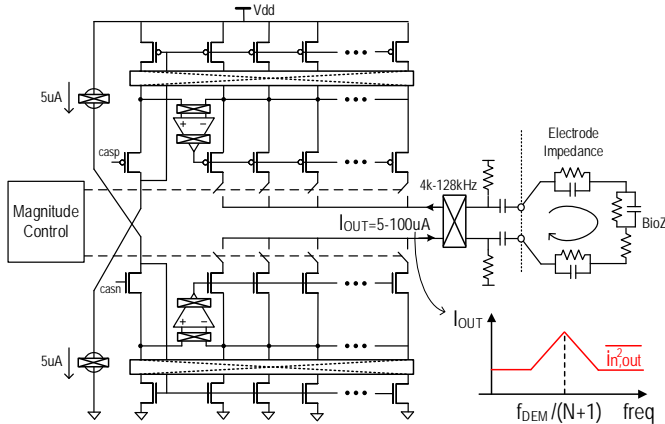


Fig. 7. Block diagram of the  $0.6\text{V}$  BioZ current source

To meet these requirements, the CS has dynamic element matching (DEM) between all unit current mirrors to modulate their  $1/f$  noise to  $f_{\text{DEM}}/(N+1)$  (Fig.7), where  $N$  is the current amplification factor. In this work,  $f_{\text{DEM}}$  is selected to be  $16\text{kHz}$  or  $32\text{kHz}$ . The CG utilizes active cascode current mirrors to improve the voltage compliance, where two OTAs regulate the  $V_{\text{ds}}$  of all mirror transistors and ensure their matching in triode region. The compliance voltage of the CG is  $400\text{mV}_{\text{pp}}$  ( $\sim 67\%$  of  $V_{\text{dd}}$ ) at the maximum current of  $100\mu\text{A}_{\text{pk}}$ . The OTAs are also chopped at  $8\text{kHz}$  to reduce their  $1/f$  noise.

## III. MEASUREMENT RESULTS

The IC is implemented in TSMC  $40\text{nm}$  CMOS and the chip area is  $1\text{mm}^2$  (Fig. 8). The readout consumes  $6.3\mu\text{A}$  from  $0.6\text{V}$  with the CS supporting current levels from  $10\mu\text{A}_{\text{pp}}$  to  $200\mu\text{A}_{\text{pp}}$ .

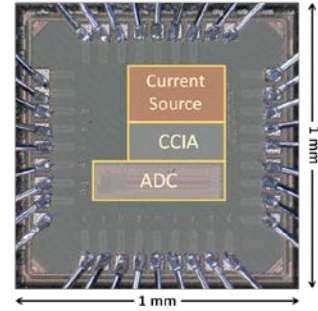


Fig. 8. Chip photograph

In Fig. 9, the ECG channel has a passband voltage gain of  $30\text{dB}$  and an input referred noise of  $145\text{nV}/\sqrt{\text{Hz}}$ , where BioZ channel is also enabled. The ECG/BioZ channel crosstalk is less than  $-60\text{dB}$  in a  $400\text{Hz}$  bandwidth. With the help of two pre-charging buffers, the CCIA's input impedance is improved by  $7\times$ , from  $20\text{M}\Omega$  to  $140\text{M}\Omega$  at  $10\text{Hz}$ . The CCIA shows its robustness to the DEO (Fig. 10). When  $300\text{mV}$  DEO is applied, the CCIA still show less than  $200\text{nV}/\sqrt{\text{Hz}}$  input noise and a flat gain of  $30\text{dB}$ .

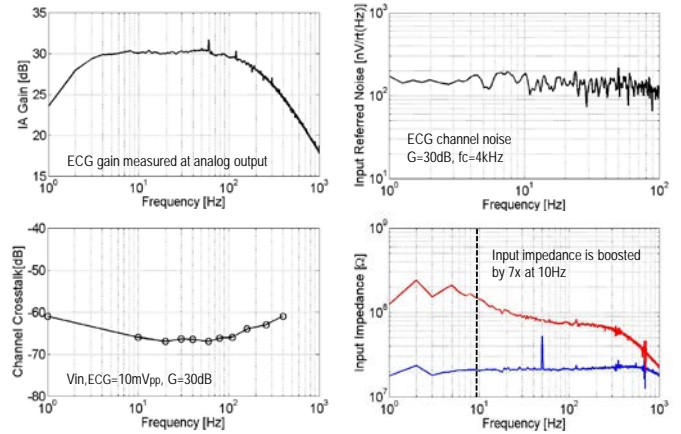


Fig. 9. ECG channel measurement results

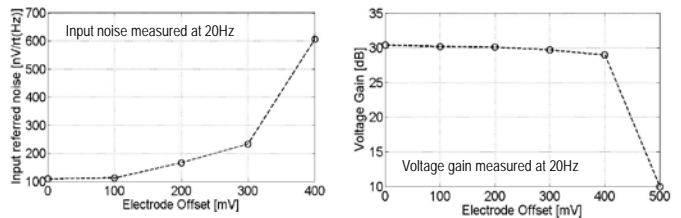


Fig. 10. ECG noise and gain versus DEO

The BioZ channel shows  $35\text{m}\Omega/\sqrt{\text{Hz}}$  sensitivity when a  $20\mu\text{A}_{\text{pp}}$ ,  $32\text{kHz}$  output current is applied to a  $100\Omega$  test resistor. This sensitivity includes the both the noise of the CCIA and the CS. Another BioZ test with resistors of  $10\text{-}200\Omega$  shows good linearity and matching with respect to theoretical numbers.

The  $13\text{b}$  SAR ADC achieves a  $9.7\text{b}$  ENOB at  $20\text{ksp}$ s while consuming  $400\text{nA}$  from  $0.6\text{V}$ . These correspond to a FoM of  $15\text{fJ}/\text{conversion}$ . The ADC power dissipation linearly increase with its sampling rate, the current is  $10\mu\text{A}$  at  $400\text{ksp}$ s.

TABLE I:

Parameters	[1]	[2]	[3]	[4]	[8]	[5]	[6]	This Work
Acquisition modes	ECG	ECG	ECG	LFP	ECG	ECG, BioZ (two IAs)	ECG, BioZ (single IA)	ECG, BioZ (single IA)
Technology	180nm	40nm	65nm	65nm	40nm	180nm	180nm	40nm
Supply voltage	0.6V	0.6V	0.6V	0.5V	1.2V	1.2V	1.8V	0.6V
Max. EDO	rail-to-rail	150mV	rail-to-rail	50mV	N/A	400mV	rail-to-rail	300mV
Input CM range	N/A	N/A	N/A	N/A	N/A	650mV <sub>pp</sub>	N/A	400mV <sub>pp</sub>
Input noise (150Hz BW)	3.44μV <sub>rms</sub>	7.8μV <sub>rms</sub>	26μV <sub>rms</sub>	4.3μV <sub>rms</sub> (300Hz BW)	1.8μV <sub>rms</sub>	0.61μV <sub>rms</sub>	0.6μV <sub>rms</sub>	1.85μV <sub>rms</sub>
Gain	34.5dB	N/A	32dB	32dB	25.7dB	28/36dB	4/16/56dB	20/30dB
Input Impedance	N/A	50MΩ	N/A	N/A	1.6GΩ@1Hz	500MΩ@50Hz	10MΩ	140MΩ@10Hz
CMRR	70dB	60dB	60dB	75dB	N/A	110dB	60dB	87dB
Power (excl.CS)	1.15μW (ECG)	3.3μW (ECG)	0.003μW (ECG)	5.04μW (LFP+Spike)	2.8μW (ECG)	56μW(ECG) 58μW(BioZ)	155μW (ECG+BioZ)	3.8μW (ECG+BioZ)
BioZ sensitivity	--	--	--	--	--	9.8mΩ/√Hz (excl.CS noise)	100mΩ <sub>pp</sub>	35mΩ/√Hz
ADC ENOB	9b	N/A	9.2b	9b	--	13.5b	10.5b	9.7b

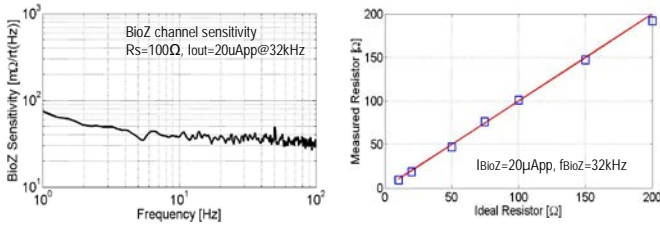


Fig. 11. BioZ channel noise and linearity

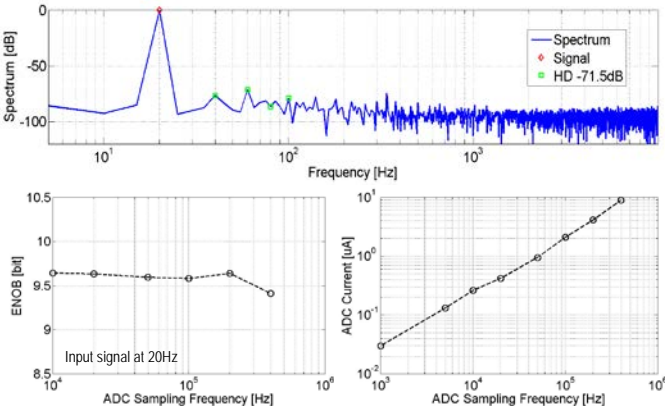


Fig. 12. ADC output spectrum, ENOB, and power

Fig. 13 shows the simultaneous ECG and BioZ recordings obtained from the same sensing electrodes on the chest. ECG signals and respiratory impedance change are clearly visible.

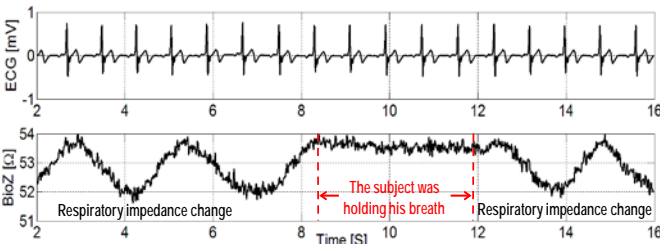


Fig. 13. Simultaneous ECG and BioZ recording from the same electrodes

Table I compares this work with prior-art low voltage ECG ICs and multimodal ECG/BioZ ICs. This work advances the existing 0.5-0.6V ICs [1]-[4] in terms of noise, CMRR, input

impedance and input CM range. Compared to multimodal ECG and BioZ ICs [5][6], this work achieves competitive accuracy but with 2x lower supply voltage and 15x lower power.

#### IV. CONCLUSIONS

This work presents the first 0.6V IC for simultaneous ECG and BioZ recording. Both signals are amplified with one single IA through orthogonal frequency modulation. The combination of a DSL and a CMFF improves the CCIA's tolerance to DEO and CM interference without adding noise. Lastly, a DEM- and active cascode-based CG realizes both low noise and improved voltage compliance.

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