A 0.6V 3.8µW ECG/Bio-Impedance Monitoring IC for Disposable Health Patch in 40nm CMOS

Jiawei Xu[1], Qiuyang Lin[2,3], Ming Ding[1], Yao Li[2], Chris Van Hoof[3], Wouter Serdijn[2], Nick Van Helleputte[3]
[1]imec - Holst Centre, Eindhoven, The Netherlands,
[2]Delft University of Technology, Delft, The Netherlands, [3]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 of an instrumentation amplifier (IA), a current source (CS) and a SAR ADC. The AFE can measure ECG and BioZ simultaneously with a single IA by employing an orthogonal chopping scheme. To ensure the IA can tolerate up to 300mV$_{pp}$ DC electrode offset and 400mV$_{pp}$ 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 140MΩ 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 35mΩ/√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 3.8µW.

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 voltage analog frontend (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 (e.g., [2][3] don't meet the noise requirement of <30µV$_{pp}$ 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 (>50µW/channel).

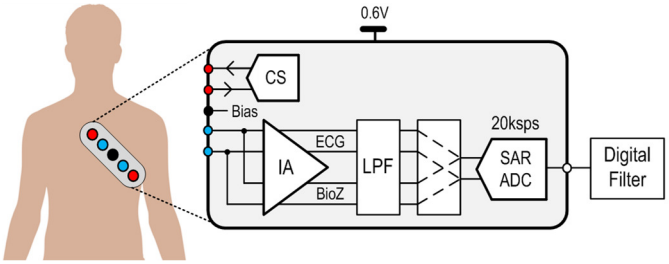

Fig. 1. IC block diagram and simultaneous ECG and BioZ measurement with one single amplifier

This work presents a 0.6V 3.8µW 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$_{pp}$ input CM fluctuation. BioZ measurement is enabled by a wide-swing and low-noise current source (CS) equipped with regulated current mirrors and dynamic element matching (DEM).

II. Circuit Implementation

A. Orthognal Frequency Modulation

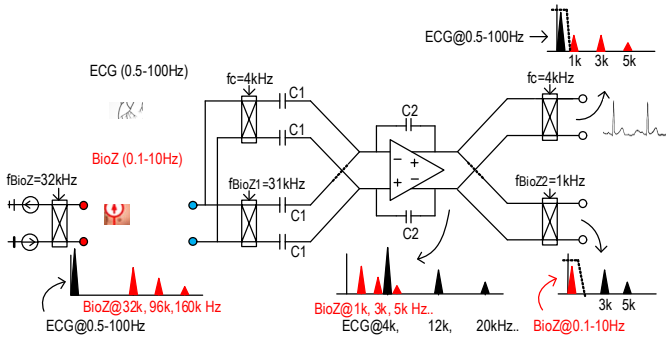

Fig. 2. Simultaneous ECG and BioZ measurement with one single amplifier

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 measures both signals from the same electrodes. The BioZ is measured by injecting a high-frequency (4k-128kHz) square-wave current. Hence, the ECG

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and modulated BioZ signals appear in different frequency bands before entering the IA. However, medically relevant BioZ spans up to 128kHz, which imposes strict BW requirements and high power consumption on the readout [6]. To overcome this issue, the modulated BioZ signal is first down-converted to an intermediate frequency (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 the ECG and BioZ signals during amplification are located at different frequency bins in an orthogonal manner. At the ECG or BioZ channel output, the desired ECG or BioZ signal is modulated back to 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 the anti-aliasing LPF in terms of its bandwidth and order. Sharing one CCIA for both ECG and BioZ channels improves system power efficiency and reduces chip area, while the orthogonal frequency modulation ensures more than 60dB signal isolation between two channels.

B. Instrumenation 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 the ECG output is tracked by a Gm-C integrator and a compensation current at fc=4kHz 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.

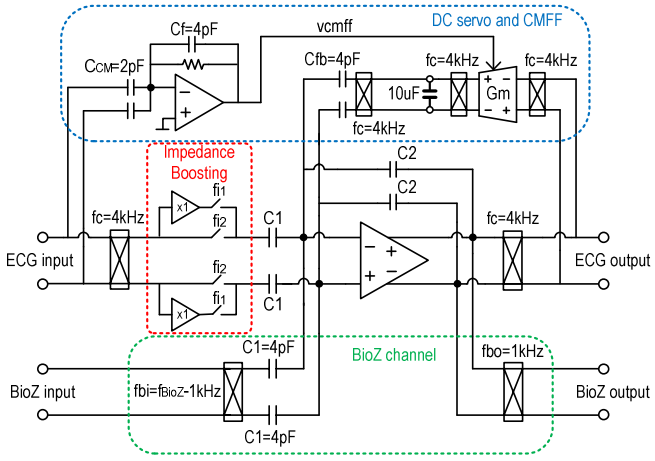


Fig. 3. Block diagram of the 0.6V ECG/BioZ instrumentation amplifier

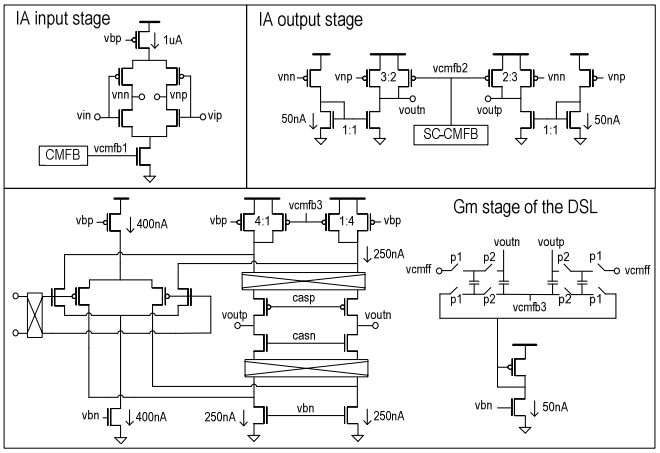
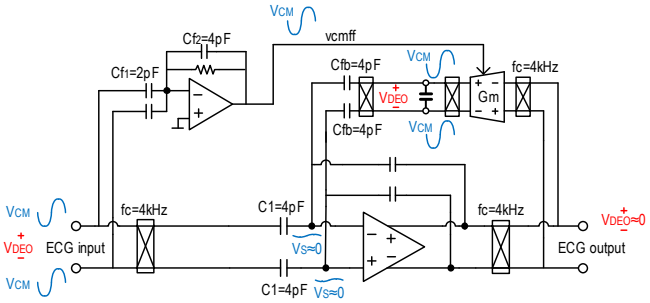


Fig. 4. Schematic of IA including inverter-based input, class A/AB output, Gm stage for DSL and CMFF

Although inverter-based input stages are attractive for low-power operation, they suffer from distortion in 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., vcmff 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 CCIAs 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 15.625µs whenever the chopping clock switches. Hence, the spike current to charge C_1 is provided by the buffers, instead of the 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 risk of instability. Thanks to the duty-cycling buffers, their noise contribution is negligible.

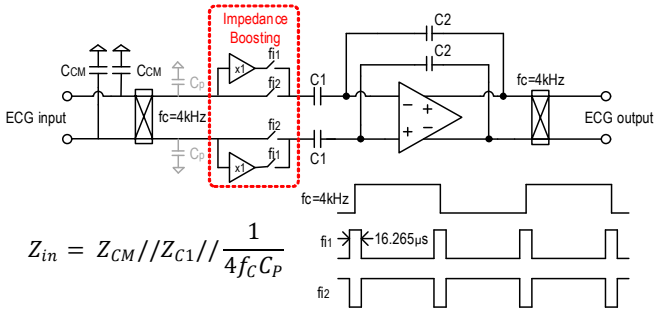


Fig. 6. Input impedance boosting with auxiliary buffers

C. BioZ Current Source

The BioZ current source (CS) is shown in Fig. 7. It has an output magnitude of 5µ-100µA and an output frequency of 4k-128kHz. Since most BioZ activities (e.g., respiration, body fluid volume) are below 10Hz, a main design challenge is to reduce the CS's 1/*f* noise for improved sensitivity. Apart from noise, achieving a large compliance range under 0.6V 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.

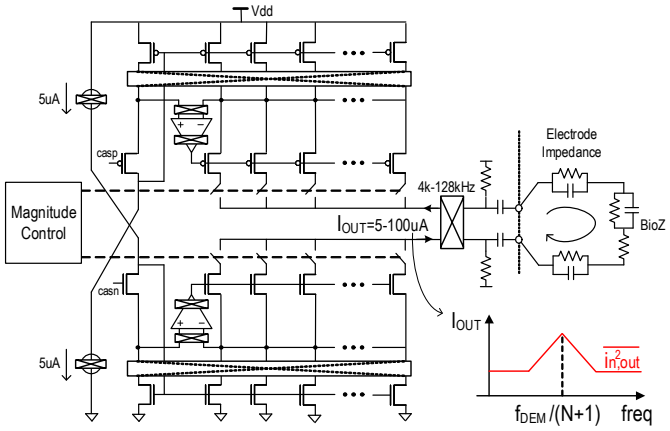


Fig. 7. Block diagram of the 0.6V BioZ current source

To meet these requirements, the CS applies dynamic element matching (DEM) between all unit current mirrors. While DEM is mostly known for its use to improve matching, it has a similar effect on the 1/*f* noise of the devices, which will be modulated, just like the mismatch effects, to $f_{DEM}/(N+1)$ (Fig. 7), where N is the current amplification factor. In this work, f_{DEM} is selected to be 16kHz or 32kHz. The CG utilizes active cascode current mirrors to improve their voltage compliance, where two OTAs regulate the V_{ds} of all mirror transistors to ensure their matching in triode region. The compliance voltage of the CG is $400mV_{pp}$ (~67% of V_{dd}) at the maximum current of $100\mu A_{pk}$ The OTAs are also chopped at 8kHz to reduce their 1/*f* noise.

III. Measurement Results

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

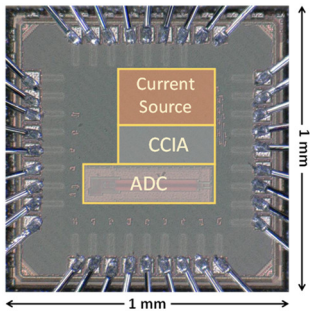


Fig. 8. Chip photograph

In Fig. 9, the ECG channel has a passband voltage gain of 30dB and input referred noise of 145nV/sqrt(Hz), where the BioZ channel is also enabled. The ECG/BioZ channel crosstalk is less than -60dB in a 400Hz bandwidth. With the help of two pre-charging buffers, the CCIA's input impedance is improved by 7x, from 20MΩ to 140MΩ at 10Hz. The CCIA also shows its robustness to the DEO (Fig. 10). When 300mV DEO is applied, the CCIA still have less than 240nV/sqrt(Hz) input noise and a flat gain of 30dB.

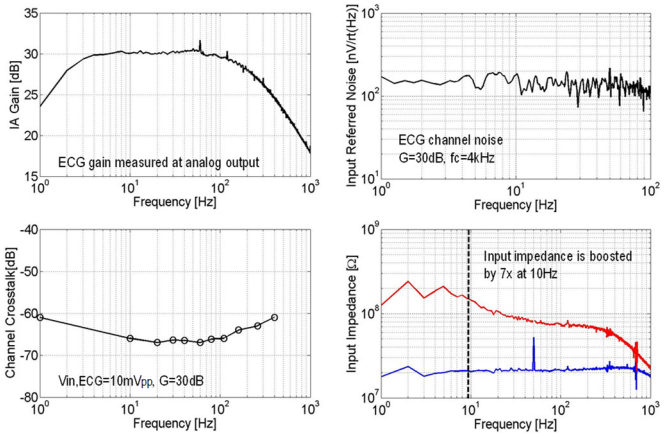


Fig. 9. ECG channel measurement results

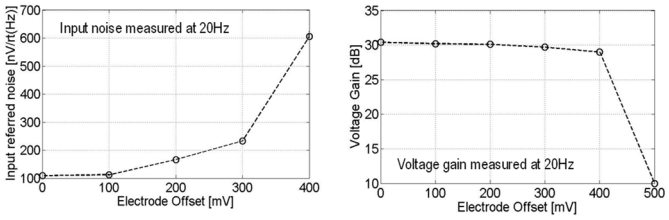


Fig. 10. ECG noise and gain versus DEO

The BioZ channel shows 35mΩ/sqrt(Hz) sensitivity when a $20\mu A_{pp}$, 32kHz current is applied to a 100Ω test resistor. This sensitivity includes both noise of the CCIA and the CS. Another BioZ test with multiple resistors between 10-200Ω shows good linearity and matching with respect to the theoretical numbers.

The 13b SAR ADC achieves a 9.7b ENOB at 20ksps while consuming 400nA from 0.6V. These correspond to a FoM of 15fJ/conversion. The ADC power dissipation linearly increases with its sampling rate. The current is 10µA at 400ksps.

TABLE I: COMPARISON TABLE OF THE ECG/BIOZ ICS

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	$650mV_{pp}$	N/A	$400mV_{pp}$
Input noise (150Hz BW)	$3.44\mu V_{rms}$	$7.8\mu V_{rms}$	$26\mu V_{rms}$	$4.3\mu V_{rms}$ (300Hz BW)	$1.8\mu V_{rms}$	$0.61\mu V_{rms}$	$0.6\mu V_{rms}$	$1.85\mu V_{rms}$
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)	$100m\Omega_{pp}$	35mΩ/√Hz
ADC ENOB	9b	N/A	9.2b	9b	--	13.5b	10.5b	9.7b

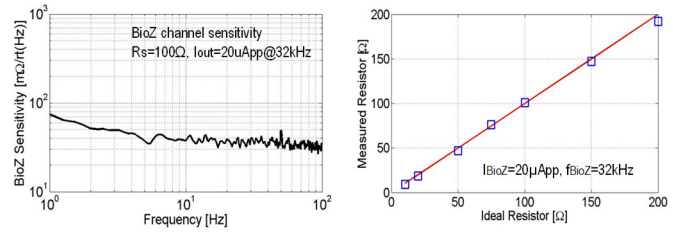


Fig. 11. BioZ channel noise and linearity

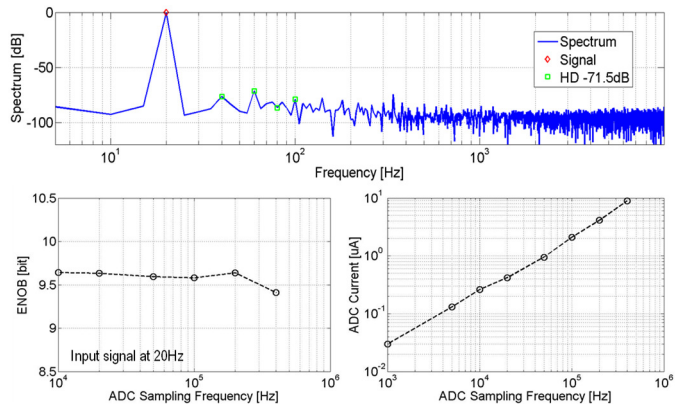


Fig. 12. ADC output spectrum, ENOB, and power

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.

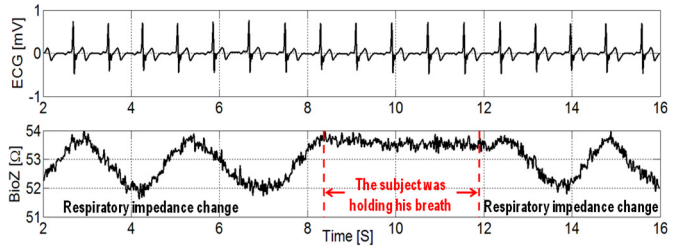


Fig. 13. Simultaneous ECG and BioZ recording from the same electrodes

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 at least 2x lower supply voltage and 15x lower power.

IV. CONCLUSIONS

The first 0.6V IC for simultaneous ECG and BioZ recording is presented. Both signals are amplified with the same electrodes and one single IA through an orthogonal frequency modulation. The combination of a DSL and a CMFF improves the CCIA's tolerance to the 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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