A 400GΩ input-impedance, 220mV$_{pp}$ linear-input-range, 2.8V$_{pp}$ CM-interference -tolerant active electrode for non-contact capacitively coupled ECG acquisition

Mingyi Chen[1,2], Ivan Dario Castro[1,3], Qiuyang Lin[1,3], Tom Torfs[1], Filip Tavernier[3], Chris Van Hoof[1,3] and Nick Van Helleputte[1]
[1]imec, Leuven, Belgium,[2]Shanghai Jiaotong University, Shanghai, China,[3]KU Leuven, Leuven, Belgium
mingyi.chen@imec.be,mychen@sjtu.edu.cn,nick.vanhelleputte@imec.be

Abstract

This paper presents a purely capacitively coupled active electrode (AE) for ECG acquisition for implementation in smart clothing or furniture. The AE has a DC input impedance of 400GΩ and 3.7μV$_{rms}$ (0.5–100Hz) input-referred noise. The whole design supports 220mV$_{pp}$ differential linear input range required in practical scenarios. An active driven-right-leg (DRL) loop is provided, achieving at least 70dB CMRR for up to 2.8V$_{pp}$ common-mode (CM) interference. Implemented in a 0.18μm CMOS process, the prototype occupies an area of 1.23mm^2, and consumes 18μA and 13μA for the AE and DRL respectively. Through-clothing non-contact capacitively coupled ECG acquisition is demonstrated.

Keywords — high input impedance, high dynamic range, common-mode interference, non-contact, capacitively coupled, ECG, active electrode.

Introduction

While electrocardiogram (ECG) is widely recognized as one of the most important physiological vital signs, existing measurement methods require a physical contact to the body which poses a large burden for widespread longitudinal monitoring for disease prevention purposes. Non-contact vital signs recording embedded in smart clothing, furniture or car seats for example, could overcome this issue. Capacitively coupled active electrodes (AEs) can acquire ECG without the need for direct skin contact, enabling unobtrusive physiological monitoring. The main principle is shown in Fig.1. One plate of the coupling capacitance is formed by the human body while the other plate can be integrated in clothing or furniture (i.e. a car seat or a bed). The coupling capacitance, typically ranging from 10pF to 200nF in such scenarios, represents a large and varying source impedance (i.e. up to GΩ-range for signals <10Hz) requiring much larger input impedance (i.e. at least several GΩ within the ECG bandwidth) than traditional contact-based recording. Furthermore, due to the large input impedance, non-contact recording is much more susceptible to (mains) interference and motion artifacts. Common-mode (CM) interference can be several volts while differential artifacts can easily be hundreds of millivolts, requiring high CMRR and large linear input range.

AEs – local amplifiers close to the non-contact electrode plates with active shielding – provide the best signal integrity [1]. A chopper amplifier with duty-cycled resistors and auxiliary-path pre-charging was presented in [2], achieving 1.6GΩ DC input impedance, still a bit low for non-contact sensing. A bootstrapped unity-gain buffer achieving 50TΩ DC input impedance was shown in [3]. However, this may exhibit a large settling time, causing slow recovery from motion or friction induced artifacts. By introducing pseudo-resistor biasing and self-calibrated positive feedback, 50GΩ input impedance at 50Hz was achieved in [4]. However, the leakage of pseudo-resistors may lead to large input offset whereas the low supply voltage and small input range will limit the ability of dealing with large interference and artifacts expected in real-life scenarios.

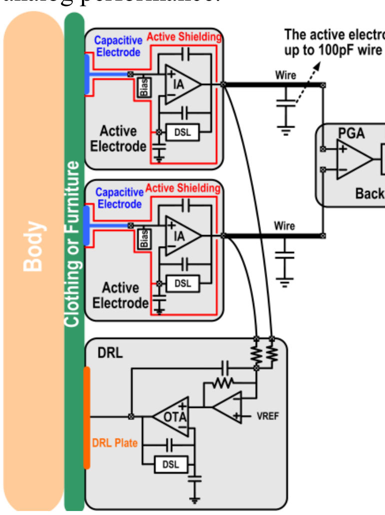
Non-contact Capacitively Coupled AE

Fig.1 shows the proposed system consisting of 3 capacitive electrodes (2-AEs and 1-DRL electrode), each with an integrated IC which can be embedded in clothing or furniture. Each AE utilizes a non-inverting AC-coupled instrumentation amplifier (IA) with a DC-servo loop (DSL). To achieve an ultra-high input impedance and fast settling time, several techniques are employed. First, the input is biased to VREF via back-to-back diodes (D_1 and D_2), with over 400GΩ DC impedance. This non-linear biasing quickly recovers to a voltage close to VREF after a large input artifact. Secondly, the custom ESD diodes at the inputs are bootstrapped so that their leakage current and parasitic capacitances are minimized. Finally, a positive feedback capacitor (C_{FB}) is used to bootstrap parasitic input capacitance (including from the package and PCB). The DSL realizes a HPF with a corner higher than the one formed by the input bias. Hence the DSL removes amplifier offset, but also allows amplification while the input is still settling to VREF after a large artifact. An extra benefit of the DSL is that the active shielding plate can be driven by the output of the DSL without an additional unity-gain buffer. Source impedance and AE gain mismatch due to die-to-die variations typically limit the CMRR significantly. Hence a capacitively coupled DRL is used to boost the CMRR.

Fig. 2 shows the block diagrams of the AE and DRL. In order to achieve a tunable sub-0.5Hz HPF corner, a pseudo-resistor (MR) is used in the DSL. Its gate voltage is derived from the source voltage of a native NMOS source follower (MN1), which can be programmable by varying the proportional-to-absolute-temperature current IPTAT. At high temperature, the V_{gs} of MN1 decreases, resulting in a first-order temperature compensation of the pseudo-resistance. A two-stage recycling folded-cascode IA is adopted to drive up to 100pF wire load (see also Fig.1). The DRL consists of a rail-to-rail input buffer followed by an AC-coupled class-AB OTA including a DSL.

Measurement Results and Conclusion

Fig. 3 shows the measured transfer function for the whole range of coupling capacitance. A stable gain and <0.26Hz HPF can be achieved for varying C_s. Fig. 4 shows the measured input impedance, with over 400GΩ DC input impedance. Fig. 5 shows the intrinsic noise of a pair of AEs measured with 200nF C_s resulting in 3.7μV$_{rms}$ (0.5–100Hz) noise. While this seems high compared to traditional ECG designs, it is important to understand that the large input impedance shaped by the C_s contributes to the total system noise. Hence the noise is highly dependent on C_s. In moderate to weakly coupled scenarios (C_s<40pF), the amplifier noise is actually not dominant. A pair of AEs has -65dB THD for differential inputs up to 220mV_{pp}, which provides a sufficiently large input dynamic range preventing saturation due to large artifacts. Fig. 6 shows the CMRR measurement results. To mimic a realistic scenario with an active DRL, the AE inputs were shorted and the 2.8V_{pp} CM signal was coupled to the inputs via a 150pF coupling capacitance. Without an active DRL, the CMRR is indeed limited because of die-to-die mismatch, but it increases to >70dB when the DRL is activated. Fig. 7 shows the bandpass filtered ECG obtained in true non-contact mode (through 2 layers of clothing). More importantly, Fig. 7 also shows the recorded ECG with induced artifacts (moving the body back and forth). Thanks to the ultra-high input impedance, high input dynamic range, and an active DRL, the ECG remains clearly visible in the recording. The 1.23mm^2 ASIC was implemented in 0.18µm CMOS technology (Fig. 8). Table I summarizes the analog performance.



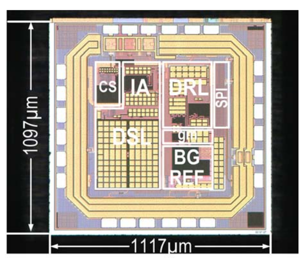


Fig. 8. Photograph of ASIC.

Fig. 1. System architectures of AE and DRL.

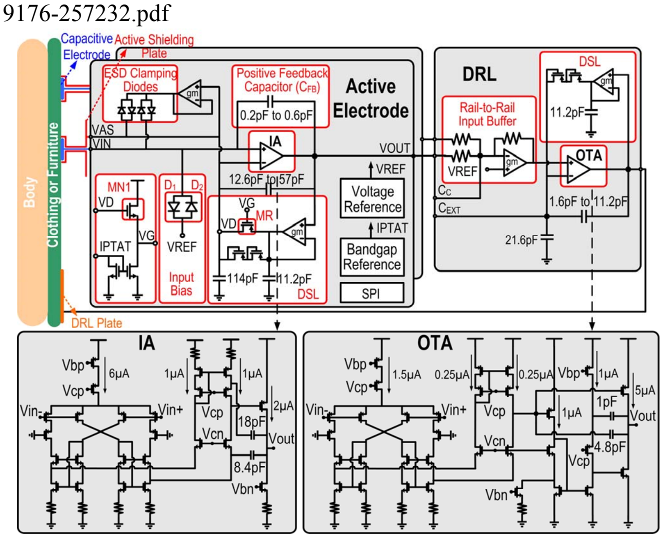


Fig. 2. Block diagrams of AE and DRL.

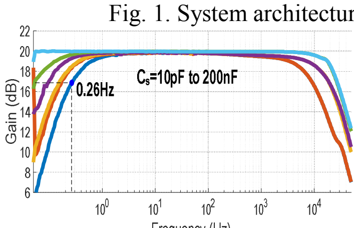


Fig. 3. Measurements of transfer function.

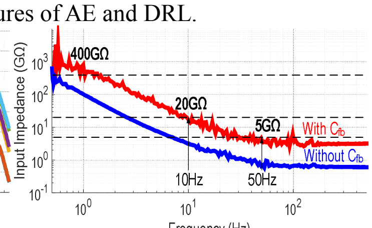


Fig. 4. Measurements of input impedance.

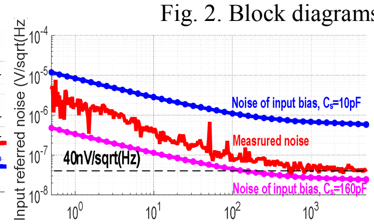


Fig. 5. Measurements of noise.

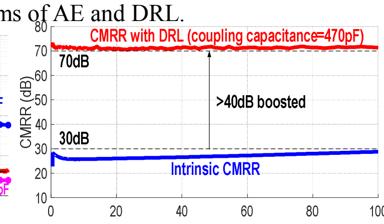


Fig. 6. Measurements of CMRR.

Table I: Performance comparison table.

Parameters		[3] JEMSTC'11	[1] ISSCC'14	[5] VLSI'16	[2] ISSCC'17	[4] CICC'17	[6] Plessey	This work	
Electrode Type		Capacitive	Dry	Dry	Capacitive	Capacitive	Capacitive	**Capacitive**	
Supply (V)		3.3	1.8	1.8	1.2	0.8	4.8-11	**5**	
Supply current (µA)		4.8	58	60	2.3	0.32	2000	**AE** **18**	**DRL** **13**
Midband Gain (dB)		0	37	0	25.7	34	34	**0/9/13/20**	
Input Impedance (GΩ)	@DC	50000[a]	1	18	1.6	200	20	**400**	
	@50Hz	---	---	6.7	1.6	50	---	**5.0**	
CMRR (dB)		---	78	108	---	66	---	**>70**	
Maximum CM (V_{pp}) interference tolerance		---	---	---	0.65	---	---	**>2.8**	
Linear Input Range (mV_{pp})		---	<51[b]	---	80	<32[c]	---	**220**	
Input referred noise/channel (µV_{rms})		---	0.65 (0.5-100Hz)	0.62 (0.5-100Hz)	1.8 (1-200Hz)	8.26 (1-400Hz)	---	**3.7 (0.5-100Hz)**	
With DRL		No	No	Yes	No	No	No	**Yes**	
Area (mm^2)		---	15.75	5.01	0.069	0.58	---	**1.23**	
Technology		0.5µm	0.18µm	0.18µm	40nm	0.18µm	---	**0.18µm**	

[a]This is the calculated value in [3].
[b][c]Calculated according to the supply voltage (1.8V[b],0.8V[c]) and the voltage gain (37dB[b],34dB[c]).

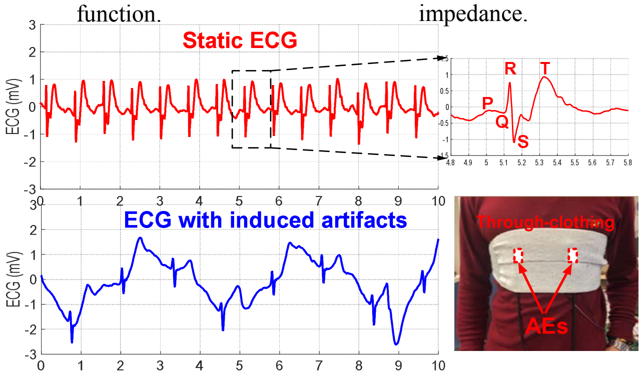


Fig. 7. Measurements of through-clothing non-contact capacitively coupled ECG.

References

[1] J. Xu, B. Busze1, H. Kim, K. Makinwa, C. V. Hoof, and R. F. Yazicioglu1, "A 60nV/√Hz 15-Channel Digital Active Electrode System for Portable Biopotential Signal Acquisition," ISSCC, pp. 96-98, Feb. 2014.

[2] H.Chandrakumar, D. Marković, "A 2.8µW 80mVpp-Linear-Input-Range 1.6GΩ-Input Impedance Bio-Signal Chopper Amplifier Tolerant to Common-Mode Interference up to 650mVpp", ISSCC, pp. 448- 350, Feb. 2017.

[3] Y. M. Chi, C. Maier, and G. Cauwenberghs, "Ultra-High Input Impedance, Low Noise Integrated Amplifier for Noncontact Biopotential Sensing," IEEE J. on Emerging and Selected Topics in Circuits and Systems, vol.1, pp. 526-535, Dec. 2011.

[4] J. Lee, H. Kim, and S. Cho, "A 255nW ultra-high input impedance analog front-end for non-contact ECG monitoring", CICC, pp. 1-4, May 2017.

[5] X. Zhou, Q. Li, S. Kilsgaard, F. Moradi1, S. L. Kappel, and P. Kidmose1, "A wearable ear-EEG recording system based on dry-contact active electrodes," VLSI-Circuits, pp. 1-2, Jun. 2016.

[6] PS25451 datasheet, Plessey, https:// www. mouser.com/ds/2/613/ 9176-257232.pdf