Surface-Mounted Dry Electrode and Analog-Front-End Systems for Physiological Signal Measurements

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Abstract—In this paper, we present a new analog-front-end system including surface-mounted dry electrode by MEMS dry electrode (MDE) and instrumentation amplifier by chopper-stabilized differential difference amplifier (CHDDA) for physiological signal recording applications. Comparing to traditional electrodes, proposed MDE shows its superior advantages for low electrode skin interface impedance and need not to use electrolytic gel. Moreover, the measured results of the presented CHDDA demonstrated the low power, denoise capability for physiological signal recording. The full AFE system is 4.5*7cm^2 in size, weight 35g and can be operated for 200Hr by 2 AAA batteries. Finally, the actual 4 channel AFE recording of EEG, EOG, ECG and EMG proves the practicability of the proposed system.

I. INTRODUCTION

Biomedical signal monitoring systems have been rapidly advanced with electronic technologies in recent years. Brain Computer Interface system that can acquire physiological signals plays an important role in real-time human physiology monitoring as well as cognitive states. Biopotential electrodes are commonly employed to transform the signals from skin tissue to the amplifier circuit and produce electrode/skin interface impedance. The electrodes placed onto the skin, say surface-mounted electrode, actually act as a voltage divider with the amplifier input resistance for signal transportation. Thus, high electrode skin interface impedance will contribute to the thermal noise and signal attenuation to the system. To lower down the interface impedance, traditional electrodes always need skin preparations, that is, abrasion of *stratum corneum* (*SC*, the outer layer of skin with electrical isolation property) and the use of electrolytic gel. However, improper skin preparation might cause skin irritation, pain, or even infection. Using electrolytic gel is uncomfortable, inconvenience and can cause itchy feeling, red skin and swollen. During chronic measurement, the conductivity of electrolytic gel will decrease gradually due to the hardened of the electrolytic gel, resulting in the degradation in the quality of data acquisition.

In this paper, the MEMS-based dry electrodes (MDE) are used as sensory unit for EEG signal acquisition. The MDE with microprobe arrays are placed onto the skin, which can penetrate SC layer into the electrically conducting tissue layer *stratum germinativum* (*SG*, the skin layer which under *SC* with electrical conduction property), but not reach the subcutaneous tissue layer so as to avoid pain or bleeding. Thus, it can avoid the high impedance SC layer and provide superior signal transmission performance than traditional electrodes that use electric gel [1]. Since MDE is expected to circumvent the high impedance characteristics of the SC, thus, skin preparation and electrolytic gel application are not required.

Traditionally, the collected weak physiological signals have to be amplified, filtered and conditioned for the actual clinical applications because the raw signals are so small that they may be strongly interfered by the environmental noise. Also, the demands of small portable, wearable and mobile biopotential sensing system are increased rapidly for the long term monitoring requirements. Thus, a low-power, low-noise amplifier system for these biopotential applications is strongly demand. To meet device miniaturization and low-power, low-noise system requirements for the modern biomedical measurement, a chopper-stabilized differential difference instrumentation amplifier (CHDDA) for MDE and DS-MDE physiological sensor applications is developed. Additionally, traditional instrumentation amplifiers suffer from the requirement of perfect resistors or current mirror match. However, resistor mismatch in the DDA design will solely influence the amplification gain. Moreover, the chopper-stabilized technique is well-known for its low-frequency noise reduction capability. Thus, the CHDDA design is used for the present low-frequency band physiological signals.

In Section II, we describe the overview of the proposed physiology measurement system. In Section III, we introduce the design methodology if the MDE and CHDDA with its superior properties in biopotential measurements. In Section IV we present experimental results and in section V we display the practical physiological signal measurements. We conclude the paper in Section VI.

II. SYSTEM OVERVIEW

Fig.1 illustrates the proposed system overview. The surface-mounted sensor MDE was placed onto the skin, and the collected raw signals were wire transmitted to the 4-channel CHDDA instrumentation amplifier system. Due to the fact that the measure setups of EEG, ECG and ECG are
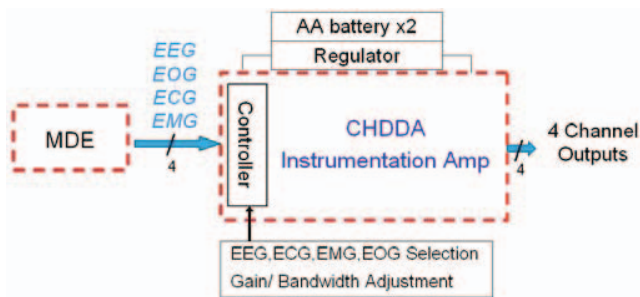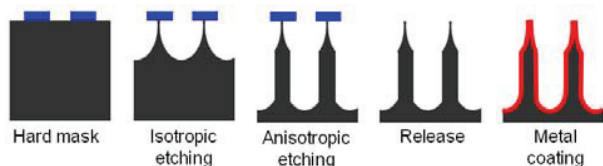
Fig. 1 Overall system structure

Fig. 2 MDE fabrication process

different, thus, a manual switch is used to select the measurement condition. Moreover, gain and filter band are also adjustable to fit the varied recording requirement.

III. DESIGN AND METHODOLOGY

A. MEMS Dry Electrode (MDE)

To fabricate the MDE sensor with probe array, a silicon microfabrication process that consisted of two-stage isotropic and anisotropic reactive ion etching with inductive coupled plasma (RIE-ICP) etching process and electroplating technology were developed and illustrated in Fig. 2: (a) circular photoresist dots to provide etching hard mask for the cylindrical probes. 6μm thick photoresist was chosen as a protection hard mask for the two-stage isotropic/anisotropic etching processes; (b) the isotropic etching process for the probe tip; (c) the anisotropic etching process for the high aspect ratio probe shaft; (d) a wet-etching process is then used to release the hard mask at the probe tip; (e) For electric conductivity, the probes were coated with Titanium/Platinum using DC-sputtering technique for high conductivity and biomedical capability

B. Instrumentation Amplifier (IA)

Current methods of instrumentation amplifier (IA) designs include traditional three operational amplifiers structure (3OIA) [2], current-balance structure (CBIA) [3][4] and differential difference amplifier (DDA) [5]. The resistor mismatch in 3OIA structure and current-mirror mismatch in CBIA can cause CMRR debasement [6] [7]. However, the variation of resistors which employed in DDA structure will only affects the gain. In this paper, we present a 1.8V DDA structure combined with chopper-stabilized technique to realize the low-power, low low-frequency noise and high CMRR for portable instrumentation design propose.
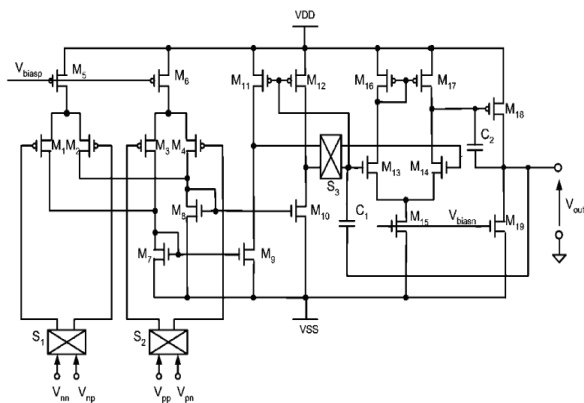

Fig. 3 Schematic of chopper-stabilized differential difference amplifier (CHDDA) [6]

The CHDDA circuit is depicted in Fig.3. The first stage is the combination of current-mirror operational amplifier with chopper-stabilized technique. The flicker noise occurs in all CMOS devices with magnitude is inverse proportional to the frequency [8]. The offset voltage occurs in the input stage due to the mismatch of the input mosfet device. In the CHDDA design, the input biopotential signals are initially modulated into high-frequency band, thus, the biopotential acts as a high frequency signal while the noise acts as a relatively low frequency signal in DDA. After amplification, demodulator was used to modulate the biopotential and noise signals back to the low-frequency and high-frequency band, respectively. Finally, a low-pass filter is used to filter out the high-frequency noise including the modulated flicker noise and offset noise. To reduce the original flicker noise, the input stage M1 and M2 are PMOS with wide width. The input stage transfer the input voltage into current signal, M7 and M8 transform the current back to the voltage signal. M9 and M10 construct a common source amplifier, M11 and M12 work as load device. M13~18 constitute a simple two-stage amplifier; C1 and C2 are Miller capacitances which increase the phase margin.

In this paper, Gm-C technique is applied for the low-pass filter design without using resistors. The second stage amplifier is formed by a simple two-stage amplifier with PMOS input transistor. After conditioning by the CHDDA and the Low-pass filter, the weak biopotential signals now will be amplified to full swing of the AFE chip.

IV. FABRICATION AND CHARACTERIZATION RESULTS

A. MEMS Dry Electrode (MDE)

The fabricated MDE were packaged by flexible printed circuit board for better handling. Fig. 4 shows the scanning electron micrographs of the fabrication results of the electrode. Etch electrode consists of a 20×20 micro probe array, etch probe is approximately 250 μm in height and 35 μm in diameter. The block bulge caused by the isotropic etching is observed at the base with an altitude about 50 μm. Thus the effective penetration length of the probe is about 200 μm.
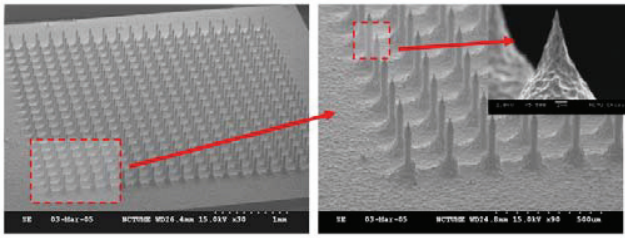
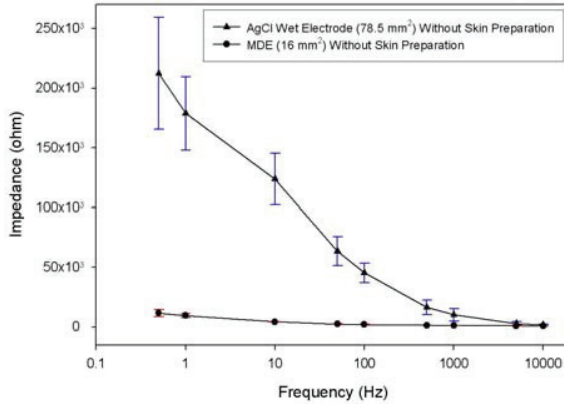Fig. 4 SEM of fabricated MDE

Fig. 5 Impedance plot of the MDE and traditional electrode
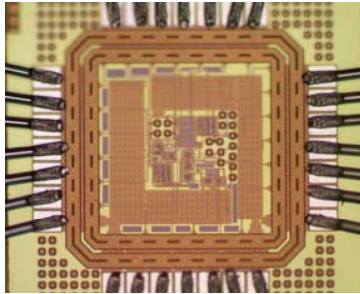

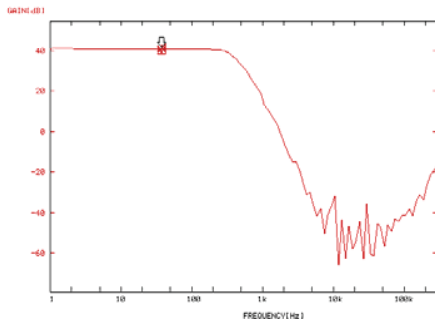Fig.6. Microphotograph of the fabricated AFE chip

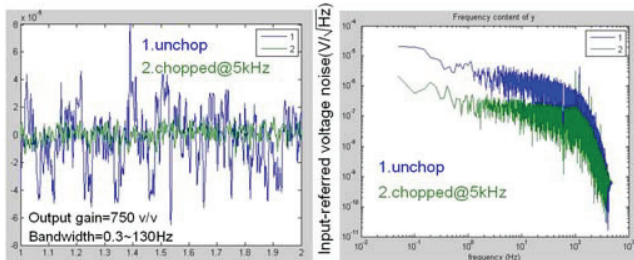Fig. 7 Frequency response of the AFE chip with 100V/V of gain

Fig. 8 Noise comparison of choppered and unchopper amplifier

Table 1
Measured AFE Chip Performance

	[1]	[2]	[3]	This work (measured)
Supply voltage(V)	1.8	±1.5	5	1.8
Total current (uA)	134	485	512	154
Power dissipation (uW)	241	1455	2560	277
Chip area(mm^2)	N/A	4.81	11.76	0.67
Mid-band gain(dB)	N/A	80	46	40 to 78
Input offset voltage(V)	N/A	60u	0.6m	40u
Input common mode range(V)	N/A	-1.5 to 1.3	X	0.2 to 1.4
Input-referred(uVrms)	N/A	0.86	2	chopped@5k=2.6
PSRR(dB)	90	65	75	67
CMRR(dB)	105	117	84	83
Full system size / weight	N/A	N/A	N/A	4.5*7cm^2 / 35g
Battery life time (system)	N/A	N/A	N/A	~200Hr (AAA*2)

The characterization of impedance of the electrode-skin interface is of utmost importance in impedance-based biosensing. Electrode impedance spectroscopy (EIS) [9] was used to evaluate the fabricated MDE. To estimate the electrode-skin interface impedance, two MDEs were attached onto skin with 4 cm distance and constructed an electrode-skin-electrode impedance architecture. The test signal was applied on the first MDE, flowed through the skin tissue and finally passed the second MDE. Therefore, the test result could be seen as the summary of two electrode-skin interface impedance. As shown in Fig. 5, the impedance plot indicates that the impedance of traditional electrode without skin preparation (but use of gel) and MDE without skin preparation at 10Hz are approximately 9kΩ and 4kΩ, respectively. The relatively low electrode/skin interface impedance proved the advantage of the proposed MDE design.

B. *Instrumentation Amplifier (IA)*

The proposed AFE chip composed of chopper-stabilized differential difference instrumentation amplifier is fabricated via TSMC 0.18um 1P6M process. Note that the ESD protection circuit was build-in the bonding pad, which is provided by the TSMC. The chip microphotograph is shown in Fig.6. The fabricated AFE chip is packaged in a dip with wire bonding, and then mounted onto a pre-designed double-side PCB as a measurement platform as described in the section II. Related voltage references are produced by regulator ICs. The chip is powered by battery; the measured overall chip current dissipation is 154□A. Measured specification results are summarized in Table 1. Moreover, the CMRR was defined and calculated as the differential gain divided by common-mode gain. The differential gain and common-mode gain were recorded to PC by DAQ card, and then calculated by MATLAB.

Fig. 7 shows the frequency response of the AFE chip with 100V/V of gain. The external capacitance is 0.01□F, cut off frequency is set as 150Hz. Fig. 8 displays the noise comparison of choppered (blue) and un-chopper (green) amplifier. Choppered amplifier shows its superior low noise performance in both time domain (left) and frequency domain (right). In Fig. 9 practical ECG Lead II recording, the upper

track and the bottom track showed the recording result with and without chopper technique, respectively. The red dotted circle denoted the ECG recording without chopper suffers from more noise comparing to the recording with chopper. This strongly demonstrated the denoise capability of the chopper stabilized amplifier.
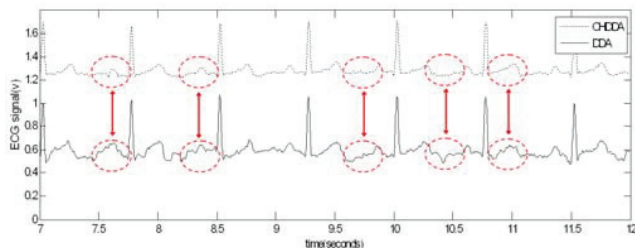

Fig. 9 The comparison of practical ECG Lead 2 recording result with and without chopper-stabilize technique

V. PHYSIOLOGICAL SIGNAL MEASUREMENT

The final analog-front-end system illustrated in section II with fabricated 4-channel CHDDA chip and MDE is demonstrated by practical physiological signal measurements in this section. Fig. 10 shows the recording results including EEG, EOG, ECG and EMG by the proposed MDE and CHDDA. Fig. 10(A) displays four channel EEG signal recorded on F5, F7, Fp1 and Fp2 from forehead. The red-dotted areas denote the EOG signal caused by the eye blinking or moving. Fig. 10(B) shows the ECG Lead I, Lead II, Lead III and V1. Finally, the recorded EMG signal from right arm biceps, right arm triceps, left arm biceps and left arm triceps are shown in Fig. 10(C).
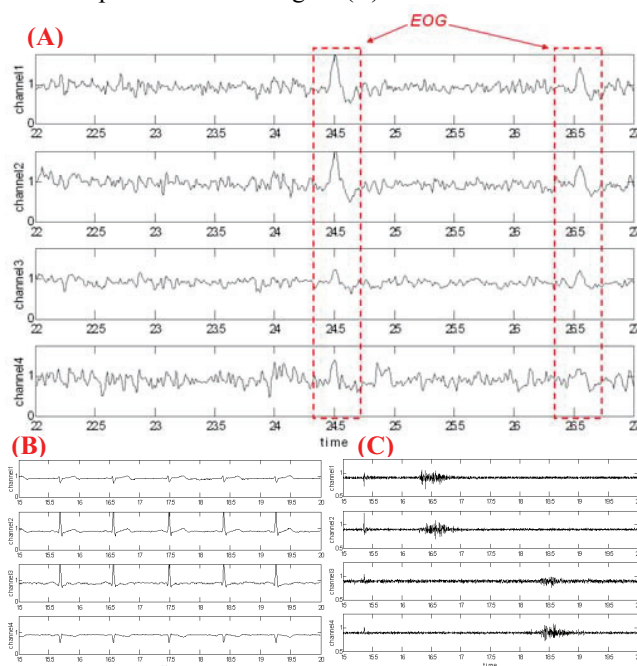

Fig. 10 Physiological signal recordings by the proposed MDE and CHDDA system: (A) EEG, the red-dotted area denotes the EOG signal caused by eye blinking or moving; (B) ECG and (C) EMG.

VI. CONCLUSIONS

In this paper, we present a new analog-front-end system including surface-mounted dry electrode by MEMS dry electrode (MDE) and instrumentation amplifier by chopper-stabilized differential difference amplifier (CHDDA) for physiological signal recording applications. Comparing to traditional electrodes, proposed MDE shows its superior advantages for low electrode skin interface impedance and need not to use electrolytic gel. Moreover, the measured results of the presented CHDDA demonstrated the low power, denoise capability for physiological signal recording. However, there is still plenty of room for performance improvements, which is still in progress. The design space is evaluated by considering the trade-offs [10]. The full AFE system is $4.5*7cm^2$ in size, weight 35g and can be operated for 200Hr by 2 AAA batteries. Finally, the actual 4 channel AFE recording of EEG, EOG, ECG and EMG proves the practicability of the proposed system.

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