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A 6.7μW CMOS Bioamplifier for Active Electrode with DC Rejection

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Abstract—Since the fabrication of active electrodes has been reported and their beneficial effects scientifically confirmed, we present a CMOS bioelectric amplifier with DC rejection designed for active electrode in this paper. Analysis, design and post simulation results will be described in detail. The amplifier operating at ±0.9V supply voltage has a mid-band gain of 40dB with ±300mV dc offset rejection and has a power consumption of 6.7μW. The bandwidth extends from a low-frequency cutoff of 7.9mHz to a high-frequency cutoff of 2.1kHz which is suitable for ECG signals. This proposed amplifier has an input-referred noise of 5.9μVrms. This amplifier is under fabrication in 0.18μm 1P6M CMOS Process.

Keywords—active electrode; CMOS; bioelectric amplifier; DC rejection

I. INTRODUCTION

Electrodes are used extensively in health care to record or monitor biopotential signals to assess various body functions, such as in measurements of EEG, EKG, EMG, and ECG. Usually a skin cleaner and a conductive paste are used to reduce the resistance and keep good conductivity between the electrode and the skin. But there are several problems with conventional electrodes with paste in use such as damage of the patient's skin, long preparation time, disability of long term measurement, etc. On the other hand, electrical interference and motion artifacts picked up by the leads may cause undesirable interference and noise problems, especially when the signals are small. To alleviate these problems a design of integrating an amplifier on the electrode for ECG measurement is applied, which is called active electrode [1]-[4]. No more skin preparations and conductive paste are required because input impedance of an active electrode is high enough compared with the skin impedance. Meanwhile, the minimum signal path between electrode and amplifier input terminal reduces main interference, while a low output impedance and a large gain on the electrode reduce the input referred noise induced by the lead wire.

The design of amplifiers for active electrode is complicated by electrode offset voltages: DC input voltage up to 300mV should not result in saturation of the amplifier. So the amplifier functions as a high-pass filter with extremely low cutoff frequency for the use of biopotential electrodes especially for high-accuracy measurement. Then a large resistance or capacitance is needed to form a large RC time constant while it leads to a slow step response. It may last several minutes before the signal is within the input range after overload for a very low high-pass cutoff frequency. Therefore we use an automatic deblocking to make the circuit response much faster. For large RC time constant, pseudo resistors are introduced to equivalent large MOS resistors but occupy a much smaller area. Besides the specifications of the amplifier described above, high input impedance, high common mode rejection ratio (CMRR), low equivalent input noise, low power consumption are also required for the active electrode application.

In this paper, a low-power and high input impedance amplifier for active electrode application is implemented. This paper is organized as follows. Section II describes the architecture of the amplifier and the practical problems of circuit implementation. Section III shows simulation results and section IV concludes the paper.

II. DESIGN

A. Proposed Amplifier Architecture

Many papers for neural recording applications use an AC-coupled capacitive feedback amplifier [5],[6] which can suppress DC-offset voltage of the electrode effectively, but it has a low input impedance compared with electrode impedance. In this paper, we present a high input impedance amplifier that input biopotential signal is connected directly to the gate of a MOS transistor.

Fig.1 shows the schematic of our amplifier for active electrode design. The amplifier functions as a mid-band filter which is based on an integrator circuit in the feedback circuit. The amplifier is unconditionally stable [7] and its transfer function $H(s)$ can be approximated as follow:

$$H(s) = \frac{sR_0C_0}{(1+sR_0C_0)(1+s\frac{A_2\tau}{A_1})(1+s\frac{\tau}{A_2})}$$

(1)

Where $A_2=(1+R_2/R_4)(1+R_4/R_3)$, $R_0$ is the equivalent impedance of pseudo resistors $M_1,M_2$; $A_1,A_2$ are DC open-
loop voltage gains of op-amps AMP1 and AMP2; $\tau_1, \tau_2$ is the time constant of dominant pole of op-amps AMP1 and AMP2.

But there are some practical problems with deblocking method. When the nMOS switch is turned on, the high-pass cutoff frequency (equivalent to a left half-plane zero of loop gain) will change to a large value which makes the circuit unstable. So we use a resistor in series with the nMOS switch to increase the zero value which can improve the amplifier's stability. When the op-amp transform to regular operating mode, actually it still remains a short time for the amplifier to be stable, but the output signal voltage has changed and the nMOS switch has turned off. A capacitance connected to the gate of nMOS switch is added to delay the time of the switch to turn off.

C. Noise

The total input-referred noise of the circuit mainly derived from op-amps, and is approximate as equation (2). Suppose we choose the same op-amp of amp1 and amp2, and its input-referred noise is $V_{n,amp1}^2$. As the resistor ratio $R_2/R_1$ is almost fixed, the total input-referred noise of the amplifier is determined by the noise component of op-amp.

$$V_{n,tot}^2 = V_{n,amp1}^2 + V_{n,amp2}^2 \left(1 + \frac{R_2}{R_1}\right)^2$$

Where $V_{n,amp1}^2$, $V_{n,amp2}^2$ is the input-referred noise of op-amp AMP1 and AMP2.

D. Op-amp

The schematic of op-amp is shown as Fig.3. Low output impedance is required because of the resistance feedback of the circuit. Meanwhile, low output impedance can reduce the noise introduced by the leads of the electrode. So a two-stage op-amp is needed because a single-stage op-amp with low output impedance isn't able to provide enough gain and has poor CMRR and PSRR. $C_C$ and $R_C$ are Miller compensation capacitance and resistance for stability. The biopotential signal we process is within the frequency of 1kHz, therefore the GBW of the op-amp has to be larger than 100kHz with a 40dB closed loop gain of the amplifier. On the other hand, the amplifier perform as a low-pass filter with a GBW/Acl (Acl is the closed loop gain) cutoff frequency. Since the amplifier works at low frequency range, flicker noise ($1/f$ noise) is a major concern for low noise application. We minimize the
effects of flicker noise by using devices with large gate areas. But devices with large areas will increase the parasitic capacitance so as to decrease the input impedance. In order to improve input impedance, the input transistor is put in a separate p-well with its substrate connected to the source. Hence, the voltage across the dominant gate to source (and bulk) input capacitance $C_{gs} + C_{gb}$ of $M_2$ remains essentially constant. To improve the bootstrapping of the source of $M_3$, the bottom current source $M_3$ is cascaded [9].

III. POST-LAYOUT SIMULATION RESULTS

The amplifier of active electrode was implemented in 0.18μm 1P6M CMOS process. A series of post simulation experiments were conducted to measure the performance of the amplifier. It can operate at a minimum supply voltage of 1.5V which is determined by the output voltage range of feedback op-amp. The measurements presented here were taken with a 1.8V supply. The reference current is 100nA. To gain a meaningful insight into the behavior of the amplifier in an active electrode, a fully CMOS integrated bias circuit for reference current of low power and small area was used [10]. Table I shows the post simulation results of the amplifier compared with other papers.

A. Frequency Response

Fig.4 shows the frequency response of the closed loop gain of the amplifier. The mid-band gain is 40dB and a high-pass cutoff frequency and a low-pass cutoff frequency are located at 7.9mHz and 2.1kHz respectively, which yield a bandwidth of 2.1kHz. Fig.5 depicts the input impedance over amplifier bandwidth range with and without impedance boosting. The input impedance with impedance boosting is 28GΩ at 100Hz while the input impedance without impedance boosting is 478MΩ at 100Hz.

B. Noise

The input-referred noise voltage spectral density of the amplifier and the op-amp is shown in Fig.6. Despite the large transistor sizes, 1/f noise still dominates. The corner frequency is typically located around a few hundred hertz and the thermal noise level equals approximately 173nV·Hz$^{-1/2}$ and 96nV·Hz$^{-1/2}$ of the amplifier and the op-amp respectively. Integrating under the curves gives noise voltages of 5.9μVrms of the amplifier and 3.4 μVrms of the op-amp from 1m to 10kHz.

C. CMRR and PSRR

Common mode rejection ratio (CMRR) and power supply rejection ratio (PSRR) was respectively measured for op-amp and shown in Fig.7. Terminals vip and vin were tied to the same input signal for measuring the CMRR and an ac coupled input signal was injected at supply terminals vdd/vss with vin
and $v_{in}$ grounded for measuring the PSRR. The CMRR and PSRR were measured 159dB and 106dB at 1 hertz respectively.

### D. Transient Response

Finally Fig.8 depicts the transient output signal. The input signal is a step voltage from zero to 300mV coupled with 5mV ac signal at 100Hz. It takes 10ms for the amplifier to be DC stable. The amplitude of output signal is 495mV.

### IV. Conclusions

A CMOS fully integrated bioamplifier featuring active DC suppression for active electrode was presented. The topology uses an integrator of a micropower op-amp within the feedback path for DC suppression and pseudo resistors to obtain a low cutoff frequency of milli hertz with a small capacitance. Deblocking method is introduced to solve the slow time-response problem. The amplifier was implemented in a SMIC 0.18μm CMOS process and was successfully post-layout simulated. It features a power consumption of 6.7μW, an input-referred noise of 5.9μVRms, a mid-band gain of 40dB, and a bandwidth of 2kHz. The amplifier integrated in the electrode is used to enhance the precision and reduce the noise of biopotential ECG signals.

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