Design of Three-stage Cascaded Two-Stage Millar Compensated OTA for Wearable ECG Sensor Technologies

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Abstract. The use of wearable sensors to monitor vital signs is an alternative to conventional medical systems in hospitals. Patients with wearable sensors can move freely while under continuous monitoring that can stably receive ECG signals, thereby improve the quality of patient care. In this work, a wearable ECG sensor is proposed. The sensor uses a Three-stage Cascaded Operational Transconductance Amplifier (OTA) as an instrument to amplify the ECG signal. The amplifier uses a Two-stage Millar Compensated amplifier integrated with N-Metal-Oxide-Semiconductor and positive-channel-Metal-Oxide-Semiconductor. The simulation results get a CMRR of 66.84 and 3.967µVrms noise and 0.039µW power and 0.981 Noise Efficiency Factor. Compared with the conventional ECG monitoring system, the entire design is smaller in size, and has low power consumption and low noise. It is observed that one can receive ECG signals stably at all times during the experiment. Also, the collected ECG signal is comparable to the ECG signal obtained using conventional adhesive electrodes.

1. Introduction
The electrocardiogram (ECG) signal is a kind of bioelectric signal that is generated by the contraction of the myocardium and can provide information on changes in the physiological function of the heart. The amplitude of ECG signal is less than 5 mV, and its frequency range is about 0.1 to 150 Hz [1].The measuring electrodes are placed in different parts of the body, and the changes in the potential difference of different body surfaces are recorded to obtain the ECG signal. Because ECG has the advantages of easy detection and better intuitiveness, it has been widely studied and applied in clinical medicine.

Ordinarily, the ECG acquisition device includes a cathode to measure the ECG signal from the human body, a simple front end (AFE) amplifier to enhance the ECG signal, a simple to advanced converter (ADC) to digitize the simple ECG signal, and a display device [2]. Heart rate variability (HRV) analysis is considered a popular method in both clinical and research fields. However, some overlooked technologies may bias their measurement results. The corner frequency of the ECG amplifier has a great influence on the RR interval detection accuracy [3].

The detection amplifier circuit is not only the gateway for the ECG signal to enter the ECG monitor, but also the premise and foundation of the ECG signal processing. Its main task is to minimize the interference signals in the ECG signal, improve the signal-to-noise ratio of the ECG signal, and provide a strong guarantee for the diagnosis of heart blood diseases. Therefore, its performance determines the performance of the entire system in a sense.

The use of advanced circuits achieves low-voltage operation, but at the same time also reduces power transmission. Since decades ago, the significance of simple circuits using low power supply voltages has been greatly expanded [4].
So far, many studies have been conducted in this field. Single-stage structure was introduced in one of the previous studies [5]. In single-stage structure, large capacitors are needed to achieve the required high gain. This results in a large silicon area on the chip, which can be a major disadvantage. Therefore, a two-stage structure was then developed, and the desired gain was achieved in two stages. Since a high cut-off frequency is set in the first stage of the two-stage structure, a large load or compensation capacitor is required at this stage. However, this leads to an increase in circuit area [6]. Now, bioelectric signal amplifiers mainly use in-phase parallel 3 operational amplifier circuits or in-phase series 2 op amp circuits [7]. In the process of ECG processing, the preamplifier has the greatest impact on the ECG signal. Therefore, it is necessary to design a preamplifier with high gain, pass bandwidth, high common mode rejection ratio and low noise for use in clinical monitoring. The buffer amplifier is used as the input stage to differentially amplify the improved signal and eliminate common-mode interference components. The general ECG amplifier uses a 6-lead ECG silicon acquisition system and uses a standard CMOS process, which reduces the design difficulty of the ECG detector and lower manufacturing costs [8]. In addition, the above paper introduces a high-power efficiency amplifier. By stacking inverters and separating the capacitor feedback network, the proposed amplifier achieves six times of current reuse, which significantly improves transconductance and reduces noise, but does not increase current consumption [9]. Nowadays, electrodes used in wearable ECG monitoring systems need to be prepared for the skin in advance, and they need to be pasted or gel to form electrical contact with the skin. In addition, since high noise spikes may appear in the data, they are not suitable for subjects with high activity levels. Ebrahim Nemati, M. Jamal Deen, and Tapas Mondal proposed a wearable ECG sensor [10]. The sensor system combines appropriate wireless protocols for data communication with capacitive ECG signal sensing and processing. The ANT protocol is used as a low data rate wireless module to reduce power consumption and sensor size. However, there is still room for improvement in terms of power and common mode rejection ratio (CMRR). This paper presents a MUlti-SEnsor biomedical IC (MUSEIC). It features a high-performance, low-power analog front-end (AFE) and fully integrated DSP. The AFE has three biopotential readouts, one bio-impedance readout, and support for general-purpose analog sensors. The biopotential readout channels can handle large differential electrode offsets (±400 mV), achieve high input impedance (>500 MΩ), low noise (620 nVrms in 150 Hz), and large CMRR (>110 dB) without relying on trimming while consuming only 31 μW/channel [11].

In order to solve the series of problems of having high CMRR, low power consumption, low noise, wearable and stable, the Three-stage Cascaded Two-Stage Millar Compensated operational transconductance amplifier (OTA) introduced in this design is an indispensable part of various portable ECG monitors. In addition to obtaining ECG signals, it also performs preliminary processing on ECG signals.

2. Method
the leftmost body circuit, imported into the fully-differential input buffer, and then the right leg drive circuit is added to the ultra-differential input buffer to make it fully-differential input buffer. The differential input buffer and the body circuit form a whole, and then enter the differential-to single-ended, and finally output the signal.

Figure 2 is the overall circuit diagram of our design. Start at the bottom left, they are the common mode power input, the voltage source of the access circuit, and the voltage source of the simulated human body circuit. Above the power input is the human body simulation circuit module, and the circuit composed of the four brothers amplifier on the right is the amplifying circuit of this design. The design uses three OTA structures as the amplifying circuit of ECG signals. The two amplifiers on the left are the first-stage amplifier circuits, which are low-pass circuits. As a fully differential input buffer, this circuit has obvious suppression of common-mode voltage gain (in ideal circumstances, Avcm is 1). At the same time, it also has a differential mode gain on the input voltage. Therefore, a very good CMRR can be obtained through theoretical calculations. The amplifier on the far right is the second-stage amplifying circuit, the output of the first-stage circuit is used as the input of the second-stage amplifying circuit, and the second-stage amplifying circuit is used as the single-ended differential of the structure. The amplifier whose input is connected to R10 and R11 is the right leg drive circuit, which is commonly used to remove the common mode signal in the input amplifier and improve CMRR; the common mode signal is connected to the human body by reverse amplifying the common mode signal to eliminate the common mode.
Figure 3. Millar Compensated OTA composed of NMOS and PMOS, a current source and a capacitor.

For the amplifier, we use the Operational Trans Conductance Amplifier that integrates PMOS and NMOS. The first advantage of CMOS devices is power saving, quiescent current consumption is very small, and the CMOS input bias current of the OP amplifier is minute as well. It has high-speed response even under low power supply voltage. In large-scale integrated circuits such as microprocessors in notebook computers and cellular phones, it can work at low voltages below 1.8V. CMOS equipment can obtain a large noise margin and is not easy to malfunction under the influence of external noise, as shown in Figure 3.

This circuit structure is an operational amplifier with MILLER compensation. I1 is an external bias current source, and M5 is a tail current source that provides DC bias for the transconductance transistor. M1 and M2 form a differential input tube. CC is a MILLER capacitor, which is used to separate the main pole and the non-dominant pole. The zero point that originally existed in the right half plane can be moved to the left half plane to stabilize the phase margin.

3. Experiments and Results

After designing the complete circuit, we simulated the circuit's differential mode output, output signal, noise. In order to simulate the human body output, we designed the human body circuit as shown in Figure 4. The x1, x2, x3 modules here consist of 2k ohm resistors in series with 1e6 resistors and 50nF capacitors in parallel to form the input and output terminals.

Figure 4. A simulated human body circuit composed of resistors, capacitors, and current sources
3.1. Bandwidth of the circuit

Figure 5. The simulation result of our differential mode gain from 100mHz to 1GHz

Figure 5 is the differential mode gain from 100mHz to 1GHz. We can see that the result is close to a straight line from 100mHz to 100Hz, and the peak value is about 41.6dB. The circuit maintains a relatively stable gain at 100Hz. The ECG signal is very weak, and the frequency range of the normal ECG signal is 0.05-100Hz. This design obtained a very stable gain in the frequency range during simulation. Next, we calculated the common mode rejection ratio of the designed circuit by dividing the differential mode gain by the common mode gain. We changed the input of the circuit to a common mode input, then analyzed the circuit according to the previous steps and performed simulations, and then found the value of the common mode gain $A_{cm}$. At this point, we get a common-mode rejection ratio of 66.84dB, which meets the requirements of the current ECG amplifier. The right leg drive circuit and shield drive circuit are used in the design, which can eliminate the common-mode voltage in the signal and improve the common-mode rejection ratio as well as the quality of the signal output.

3.2. Noise analysis

In the noise simulation of this design, we first measured the magnitude, which is 41.6dB and 121V/V with log. Then the total RMS noise is 479.92 after the simulation, and the At 479.92µVrms output is calculated using the following formula. noise, the $\text{IRN} = 3.967\mu\text{Vrms}$.

$$\text{IRN} = \frac{\text{The total RMS noise}}{\text{Noise voltage}} \quad (1)$$

In addition, we also calculated the Noise Efficiency Factor (NEF). ECG sensing is typically performed by precision analog circuitry which amplifies weak, low bandwidth bio-signals while introducing minimal circuit noise. With the power consumption being noise-limited, achieving high noise efficiency has always been an important aspect in the design of such amplifiers for physiological sensing applications. The NEF and the Power Efficiency Factor (PEF) are well-known metrics to quantify the performance of these amplifiers. Using the following formula, we get NEF

$$\text{NEF} = \frac{2I_{tot}}{V_{T}^{4}k_{B}T\pi BW'}$$

(2)
where \( V_T \) is the thermal voltage, \( k_B \) is Boltzmann's constant, \( T \) is the temperature, \( I_{tot} \) is the total current drawn by the amplifier, \( BW \) is its bandwidth, and \( v_{ni,RMS} \) is its input-referred noise. For a differential amplifier using MOSFETs as the input devices operating in subthreshold and being the sole contributors to the noise [12]. Here we take the temperature \( T \) as 300k and the bandwidth as 100. The final result is approximately equal to 0.981.

3.3. Power consumption

For power, we multiply the output current by the output voltage to get 0.039\( \mu \)W, which shows that this is a very low-power amplifier circuit. In addition, the total capacitance used by the entire amplifying circuit is 1.021nF. The size of the capacitance has an effect on the current of the starting winding. The larger the capacitance, the larger the current of the starting winding.

3.4. Brief summary and other characteristics

To sum up, the final performance of the designed amplifier is shown in Table 1. The differential mode gain is 41.6dB, the common mode rejection ratio is 66.84dB, the Total Integrated input-referred noise, Total power consumption of all blocks is about 0.039\( \mu \)W, the Total amount of employed capacitance is 1.021nF, and the Noise Efficiency Factor is 0.981.

![Table 1. Simulation parameters and Values](image)

| Simulation parameters | Values |
|-----------------------|--------|
| Differential gain (dB) | 41.6   |
| Common-mode rejection ratio (CMRR) | 66.84 |
| Integrated input-referred noise(\( \mu \)Vrms) | 3.967 |
| power consumption of all blocks(\( \mu \)W) | 0.039 |
| employed capacitance(nF) | 1.021 |
| Noise Efficiency Factor (NEF) | 0.981 |

4. Discussion

The electrocardiogram is the simplest, quickest and most economical method for judging whether our heart is healthy. In this design, we have overcome several challenges of the ECG amplifier circuit on the market. First, danger might occur due to high voltage and current when the ECG subsystem itself or other medical equipment connected to the patient or operator fails. The ultimate goal of the design is to ensure the safety of patients and operators from such voltage or current damage. Because our electrode current is very small, it is relatively safe for patients or doctors to operate. Second, the design has stable output and high gain. Third, ECG signals may be damaged by a variety of interference sources, including power line interference, contact noise between electrodes and skin, muscle contraction, and electromagnetic interference from other electronic devices. Any number of interference sources may cause the ECG baseline to shift or display electrical noise. For clinicians, the most important thing is that the ECG signal is clear and readable, and it will not cause ECG diagnosis errors. In terms of errors, the noise found in our design is extremely small, only 3.967\( \mu \)W, which is one of the best performances of the design. Because our design only uses components such as resistors and capacitors, and the amplifier space is small, easy to integrate, and wearable.

5. Conclusion

By understanding the importance of human ECG signal acquisition and the characteristics of current ECG monitors in society, we proposed the design of a Wearable ECG Sensor Technology. In this paper, we have studied the characteristics of ECG signal energy mainly concentrated in 0.05-100Hz, through Three-stage Cascaded Two-Stage Millar Compensated OTAs; as a result, an ECG amplifier circuit with ultra-low power consumption, low noise and stable gain is realized. Compared with the 12-18 lead medical-grade ECG equipment, this study is driven by the human body's 2-lead and right leg, and because the CMOS integrated amplifier has the characteristics of saving space, it achieves wearability.
With the low power consumption and stable gain of the amplifier circuit, consumers can wear it by themselves and record their own ECG activity for a long time without delaying the condition. At the same time, this is also very helpful for elderly people with healthy heart but weak legs and feet. In the future, people will pay more and more attention to health, and people are more efficient, fast and convenient, therefore, we believe that the demand for wearable ECG monitoring equipment will increase.

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