Low-Noise Electrocardiogram Analog Front-End

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Abstract. In recent years, the increasing emphasis on personal health monitoring has accelerated the development of wearable medical devices, with notable trends toward miniaturization, user-friendliness, and intelligent features. As a vital bioelectrical signal indicative of cardiovascular diseases and other medical diseases, the research of electrocardiogram (ECG) monitoring has become a critical focus. Typically characterized by low amplitude (0.1-5 mV) and narrow frequency bandwidth (0.05-100 Hz), ECG signals are inherently vulnerable to semiconductor circuit noise, particularly low-frequency flicker noise. This susceptibility underscores the importance of high-performance, low-noise analog front-end (AFE) circuits in ensuring accurate ECG signal acquisition. This paper first introduces classical AFE circuit designs briefly, including their internal components and functional characteristics. Subsequently, the paper systematically categorizes and analyzes various low-noise optimized designs for AFE implementations, with particular emphasis on flicker noise reduction strategies. Through structured exploration, it demonstrates both the technical feasibility and crucial relevance of advanced low-noise AFE designs, reinforcing their significant role in advancing ECG system performance.

Keywords: Electrocardiogram, Analog Front-End, Low Noise.

1 Introduction

In recent years, the high risk of cardiovascular disease (CVD) has imposed a considerable burden on patients and medical treatment worldwide [1]. This condition is particularly threatening to the elderly, whose physical functions are gradually deteriorating. Real-time monitoring of ECG signals can assist doctors in understanding the patient's condition quickly, making it one of the most important indicators for detecting and diagnosing cardiovascular diseases [2]. Continuous monitoring of physiological parameters can be achieved through portable health monitoring systems, such as wearable and implantable devices [3]. Figure 1 illustrates the design of a Wearable Healthcare Monitoring System (WHMS). This remote monitoring design enables healthcare providers to evaluate patients' real-time physical conditions, even when they are not in the hospital for treatment. Additionally, it not only makes timely interventions possible but also reduces the duration of medical treatment and the cost of examinations.

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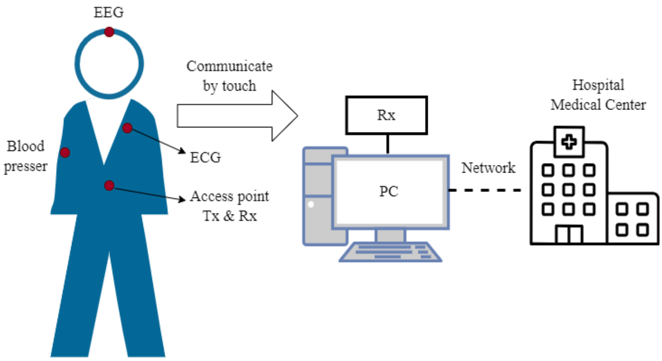

Fig. 1. Wearable healthcare monitoring system.

A wearable healthcare monitoring system consists of a detection device, a receiver such as a computer, and a medical monitoring center. The detection equipment can be placed directly on the patient's specific detection site, and when it detects the bioelectric signal, it will transmit the collected signal data remotely to the receiving end. Subsequent to data processing, the receiving end transmits the requisite data to the medical center via the network, thereby enabling efficient monitoring of patient indicators by medical professionals and equipment. The remote portable monitoring system described above relies on the amplification of signals by an analog front end (AFE). Thus, an AFE with particular characteristics must be included in a dependable, portable, real-time ECG monitoring system. For instance, the amplitude and frequency of bioelectrical signals are usually quite small, which makes them typically difficult to acquire [4]. The process of electrical signal acquisition is also affected by the inherent noise of the circuit, which makes the acquired signal less precise and more challenging to attain. Consequently, engineering an AFE with low noise and power consumption is essential.

This paper provides a brief introduction to the low-noise AFE. The first chapter contains the overall structure of an AFE circuit in a conventional design, with a more detailed description of each of its internal components and their functional characteristics. The second chapter is designed to optimize the low-noise design of multiple AFE circuits, with particular emphasis on optimizing the design for flicker noise and low-frequency noise.

2 Overall Structure of the AFE Circuit

It's evident that the human body's finger activity markers are often reflected by bio-signals with specific amplitude and frequency requirements [5]. Among them, small amplitude and low frequency are considered the most representative. In addition to meeting the requirements for acquiring bioelectric signals from the human body, the circuit within the AFE, resistors, MOS tubes and other electrical components will generate a certain amount of noise itself. These include flicker noise, thermal noise, and electrode DC offset (EDO) [6], all of which can interfere with the detection of bio-signals and reduce the accuracy of the monitoring results. Consequently, ECG analog front-end circuits should be designed to meet the performance requirements of low noise, high amplification and low power consumption [7], and also endeavour to meet the supply requirements of portable wearability and low cost [8].

In the design of classical AFE circuits, it is common to use an instrumentation amplifier (IA), a programmable gain amplifier (PGA), a low-pass filter (LPF), and an analog-to-digital converter (ADC) [9]. The architecture of a classical AFE circuit is displayed in Figure 2.

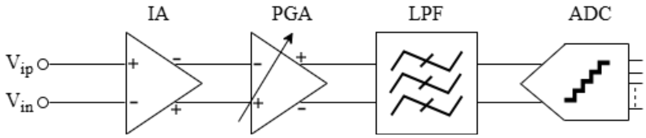

Fig. 2. The architecture of a classical AFE circuit.

The subsequent discussion will provide a comprehensive overview of the functional roles of each component within the AFE. IA is the input stage of the AFE circuit, which is responsible for the initial amplification of the input ECG signal. Simultaneously, the IA determines the common mode noise, common mode rejection ratio (CMRR), input impedance, and other important performance parameters of the AFE. During the design process, common types of IA include three-op-amp instrumentation amplifiers, current-feedback instrumentation amplifiers, and "AC-coupling-capacitor-feedback" amplifiers [10]. Three-op-amp instrumentation amplifiers have high impedance input signals through transistor gates. However, these amplifiers consume large amounts of power and have poor gain accuracy. Although current feedback instrumentation amplifiers eliminate clock burr interference, transconductance limits the input linear range and their gain accuracy is insufficient. The "AC coupled-capacitor feedback" amplifier circuit is notable for its simplicity, small size, and common usage in biological signal acquisition circuits. Nevertheless, its single-ended input and single-ended output common mode rejection capability is slightly inferior. The selection of an AFE is informed by the consideration of their distinct advantages, drawbacks, and functional characteristics.

Due to the varying amplitudes of bio-signals during the acquisition process, a programmable gain amplifier (PGA) is required in the AFE to adjust the gain, compensating for the insufficient amplification from the IA, ensuring that the ADC receives high-precision data signals [9]. Programmable gain amplifiers can be classified into two categories based on the type of feedback: resistor-feedback PGA and capacitor-negative feedback PGA. The latter is advantageous due to its ability to eliminate thermal noise while maintaining higher matching and gain accuracy, and its tendency to generate less quiescent currents and consume less power. Thus, a capacitor-type PGA is generally used in the AFE circuit of ECG to obtain bio-signals [8].

The low-pass filter is primarily used to remove out-of-band noise and interference signals captured by the AFE. It is also employed to prevent sample aliasing of the ADC. Common filters used in circuits include RC passive filters, RC active filters, switched-capacitor filters [11], and GM-C filters [12]. Nevertheless, conventional RC filters often struggle to meet the low-pass cutoff frequency requirements for bio-signal acquisition. These filters also tend to occupy excessive chip area to meet the high capacitance and resistance requirements. Switched-capacitor filters offer high linearity and favorable PVT (Process, Voltage, Temperature) characteristics; however, they introduce additional noise into the circuit, requiring greater power consumption to alleviate this noise. GM-C filters have the benefits of low power consumption and ease of integration, yet they are also characterized by limitations in input range and poor linearity. In practical applications, the source degradation technique is frequently applied to extend the linear range of the transconductance unit [6].

The ADC can further process the collected bioelectric signals through digital conversion, thereby avoiding aliasing during sampling. Given the necessity of accuracy and speed in bio-signal acquisition, the utilization of SAR ADCs [13] and Σ-ADCs [14] is prevalent. In comparison, SAR ADCs typically use dynamic comparators and contain numerous digital circuits, resulting in relatively low power consumption. Furthermore, to meet the low-frequency characteristics of ECG bio-signals in the AFE, general ADCs are often replaced with SAR ADCs based on Nyquist sampling in the circuit. On the other hand, Σ-ADCs employ oversampling and noise shaping techniques, which modulate the noise to frequencies above the signal bandwidth, thereby enhancing the resolution. This makes them suitable for low-speed, high-precision bio-signal acquisition AFE.

3 Low Noise Optimization of AFE Circuits

The optimization of AFE circuits for noise reduction constitutes a critical facet of circuit optimization. This optimization can be achieved through various approaches, including the enhancement of modules at different levels. Given the heterogeneity of noise types and their respective frequencies, it is imperative to employ bespoke optimization methods, tailored to the specific characteristics of each noise type. This necessitates the development of customized circuit designs or the refinement of existing methods to mitigate noise. This section will introduce the optimized design to reduce flicker noise and other noises.

3.1 Reduces flicker noise

The chopper stabilization technique is frequently utilized to minimize flicker noise in ECG amplifiers [15]. It utilizes square wave modulation and demodulation to modulate 1/f noise to a higher frequency before filtering out the high-frequency 1/f noise using a low-pass filter.

In 2018, Bai [16] proposed a 0.5V ultra-low-power ExG AFE circuit, which integrates a capacitive-coupled chopper instrumentation amplifier (CCIA), switched-capacitor filter (SC-Filter), and a fully dynamic 10-bit SAR ADC. It also implements techniques such as electrode DC offset elimination, chopper ripple suppression, and input impedance enhancement. For the chopper amplifier, it is used to amplify the input signal and modulate the 1/f noise from the circuit. The first stage employs a folded-cascode operational transconductance amplifier (OTA), and the second stage uses a

Class-A amplifier structure to ensure large output swing and high linearity. Moreover, the circuit incorporates a DC feedback loop (DC-Servo Loop) and ripple reduction circuit (RRL) to suppress electrode DC offset and circuit interference. The entire circuit adopts a fully differential structure, which enhances common-mode rejection and eliminates the interference of even harmonics of the signal. The article also presents a low-power DC-AC (direct current-AC) configurable ECG signal AFE interface circuit. The signal acquisition and processing path of the interface circuit consists of an IA, a PGA, a continuous time sigma-delta modulator (CTΣΔM) module, and a right-leg-driven (RLD) circuit. The RLD circuit is incorporated to eliminate the interference of the 50/60 Hz IF power line. Furthermore, the Power Management Unit (PMU) and Clock Generator circuits are integrated within the chip to optimize the overall ECG acquisition system. The power consumption of the circuit is significantly reduced by regulating the operating and sleep modes of the circuit. Nonetheless, the study indicates that the circuit design exhibits limited noise reduction for weak electrical signals. Subsequent research should focus on the investigation of AFE circuit architectures that are adapted to various environments and processes. It should be acknowledged that the scope of application in the present article is limited.

In the same year, Xu et al. [17] proposed a 1.2 V 36 μW reconfigurable analog front-end (R-AFE) as a general-purpose, low-cost IC for multimodal bioelectrical signal acquisition. The R-AFE can reduce the 1/f noise from the preamplifier and current generator (CG). The preamplifier is set up as an AC-coupled chopper IA with resistive feedback in order to accomplish this. Furthermore, ensuring that crucial parts, such as operational amplifiers (OPAs), feedback networks, and input bias networks, can be easily switched between acquisition modes without sacrificing performance is a major design issue for reconfigurable preamplifiers. Thus, to suppress differential mode interference or artifacts, the OPA should have rail-to-rail voltage headroom at both inputs and outputs. The complementary input pair and AB-class output based on cross-linear loops can fulfill these requirements. The purpose of this highly reconfigurable AFE is to capture respiratory and cardiovascular signals in a portable and compact manner. The AFE has a footprint of 1.1mm2 and supports five sensing modes: ECG, BioZ, GSR, PPG and GPA. It outperforms existing reconfigurable ICs with superior noise-to-power ratios and promotes state-of-the-art, dedicated biomedical sensing in a smaller footprint.

3.2 Reduce other noise

In addition to the design of ECG AFE circuits that can reduce flicker noise, there are also studies that employ various methods and techniques to reduce different noises.

In 2021, Zhang [6] designed an ECG acquisition AFE chip with low noise and low power. In this design, chopper modulation technology is introduced in the instrumentation amplifier to reduce the low-frequency noise of the AFE. Additionally, a chopper-capacitor-chopper integrator based DC servo loop (C3IB-DSL) is proposed, with the dual objectives of suppressing the electrode DC offset, enhancing the PVT characteristics of the CCIA, and reducing the noise introduced into the CCIA by the DSL. To further optimise performance, a low-noise, high-precision dual-supply ECG acquisition AFE chip is also proposed. The circuit includes CCIA, Gm-C LPF and continuous time sigma-delta (CT Σ-Δ) modulator modules. By combining high and low level power supplies, the circuit is optimised for noise, power dissipation, and electrode DC offset voltage cancellation range, which are key specifications. In the meantime, a common mode feed forward circuit is introduced into the CCIA to improve the common mode rejection capability of the CCIA. According to simulation results, the AFE has a large electrode DC offset cancellation range, a common mode tolerance range, low power consumption, and good noise performance. All the indices meet the index requirements of the AFE for ECG acquisition. Still, there are issues with the design that need to be resolved. Since the dynamic range of the traditional ADC is limited, the traditional analogue front-end needs to add an instrumentation amplifier before the ADC to first amplify the weak bioelectric signal and then digitise it by the ADC. Investigating the direct quantization method for bioelectric signals is the next stage. In this approach, the ADC is directly connected to the electrodes, and the bioelectric signal is quantized directly. This eliminates the need for modules like instrumentation amplifiers and programmable gain amplifiers, which are typically used in traditional analogue front-end circuits.

In 2024, Chen et al. [18] designed an amplifier that has low total harmonic distortion (THD), minimal noise, high gain, and little power consumption. This amplifier is based on chopper stabilization, capacitive feedback, and current reuse techniques. In order to enhance the power supply rejection ratio (PSRR) and CMRR, the amplifier additionally utilizes a fully differential architecture. Meanwhile, the employment of a fully differential structure is instrumental in decreasing even harmonics, consequently lowering the THD level. The use of a two-stage OTA is also applied to achieve a closed-loop gain of 59.7 dB and ensure that gain error is minimized. Furthermore, the design integrates cascade coding and current reuse techniques, enabling high gain with reduced power consumption, and further lowering the power consumption of the OTA. What's more, this design can be used not only to record ECGs but also to monitor other biomedical signals such as EEG, EMG, etc. due to its wider signal bandwidth and greater gain. Conversely, this amplifier's practical application scenarios are constrained, and their scope can be expanded through further exploration to identify viable avenues for its deployment.

4 Conclusion

This paper primarily discusses low-noise AFE circuits for ECG. Initially, the characteristics of ECG and the overall structure of AFE circuits in general-purpose designs are described in detail, with an emphasis on the roles and features of each part within the AFE circuit. Subsequently, the paper delves into the low-noise optimized design of several distinct AFE circuits. This includes the application of chopper stabilization techniques to eliminate flicker noise and the utilization of various methods to eliminate other low-frequency noise.

The human body contains not only ECG signals but also signals from other biological systems, such as electroencephalographic (EEG) and electromyographic

(EMG) signals. These signals possess distinct amplitude characteristics, noise levels, and other indicators. This paper merely provides a concise introduction to the AFE circuit for ECG measurement. Subsequent research endeavors may explore the optimization of low noise for diverse bioelectric signals with greater depth. In addition, the optimization of AFE circuits involves many other aspects, such as power consumption, bandwidth, CMRR, and so forth. Each of these metrics exerts a significant influence on the design of analog circuits. In the future, the optimization of each performance metric of analog circuits for ECGs can be studied in depth, as well as the optimization of different performance metrics for different bioelectric signals. Consequently, future research will provide a more detailed introduction to the AFE designs for various biological signals and discuss the diverse optimization strategies within the circuits to improve their performance metrics.

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