

Abstract

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Project Title : An Ultra-Low-Power Analog Front End (AFE) Design for Wearable Cardiac Monitoring Systems

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Abstract:

Cardiovascular diseases and heart failure have become the leading cause of deaths in Thailand in the past decades. To prevent deaths due to heart attacks, wearable cardiac monitoring systems can be used to detect abnormalities in the heart's electrical patterns and issue warning messages to the patients himself or to the nearby health centers for immediate assistance. To be practical, such wearable monitoring systems need to be very low power and small in size for patients' comfort, thus requiring that most of the circuitry be integrated onto a single integrated circuit. Analog front end (AFE) is one of the most crucial circuitry in the cardiac monitoring systems and other sensor interfaces since it determines the quality of the desired signal and has strong effects on the overall power consumption and size of the system. In this research, we have designed and built an AFE integrated circuit (IC) for wearable cardiac monitoring systems. The design focuses on maximizing the AFE's performance such as maximizing the open-loop gain, minimizing the input-referred noise, and minimizing the total power consumption and chip area. The IC has been fabricated in a commercial 0.18 μm CMOS technology and tested to validate its functionality. Thus far, three publications in ISI journals have come out of this research, two in International Journal of Circuit Theory and Applications, and another in The IEEE Transactions on Circuits and Systems II: Express Briefs. The results from these study will be useful not only in the design of wearable ECG acquisition systems but also in other energy-constrained applications such as implantable biomedical devices, wearable electronics for health monitoring, low-power sensor nodes in wireless sensor network, and wearable biometric devices.

Keywords : wearable cardiac-monitoring devices, front-end amplifier, biosignal acquisition, current-reuse

Final report contents:

1. Abstract

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2. Executive Summary

2.1 Introduction to Research

Cardiovascular diseases and heart failure have become the leading causes of death throughout the world in the last few decades. In some developed countries such as the USA, heart failure accounts for almost one-third of the overall death cases and is the number one cause of death in adults. In Thailand, specifically, the rate at which the Thai population is developing cardiovascular diseases is increasing at an alarming rate due to the change in lifestyles such as consuming diets high in salt and saturated fat, lack of exercise, smoking and drinking, and leading stressful lives, etc. Moreover, recent advances in medicine have helped extend the human's lifespan, which result in a growing number of elderly people in the Thai society. These aging people possess

a higher risk of acquiring cardiovascular diseases and heart failure. These non-communicable diseases have become an epidemic in the modern Thai society and have claimed many lives of our fellow Thai citizens yearly. According to the 2008's statistics from the World Health Organization (WHO), cardiovascular diseases account for 27 percent of all deaths of the Thai population—by far the leading cause of deaths in our country (www.who.int/countries/tha/en/).

It is well known that, for patients with a high risk of heart failure, deaths can be avoided if the patients receive proper immediate treatments. Now imagine if these patients possess wearable cardiac monitoring systems that can alert them when their hearts are about to malfunction; the system can sense the heart's electrical activities called electrocardiograms (ECG) through the electrodes placed on the patients' skin; it then processes the recorded signals to determine the abnormalities in the ECG's patterns called cardiac arrhythmias; once the arrhythmia is detected, depending on the degree of seriousness, the system can deliver warning messages to the patients themselves, to their relatives, or to the medical personnel at the nearest hospital for emergency assistance via a communication network. Such cardiac monitoring system can help improve the patients' chances of survival if helps arrive in time.

In this research, we have designed, built, and tested an ultra-low-power low-noise analog front end (AFE) intended for use in such wearable cardiac monitoring systems. Even though this research focuses on the design of an AFE for cardiac monitoring applications, the knowledge gained will also be very applicable to other biosignal acquisition systems and other systems that need to amplify tiny signals with stringent power, noise, and size requirements.

2.2 Literature Review

Research on the designs of AFEs for biosignal acquisition is currently one of the hottest topics in the field of biomedical circuit and system designs. This is due to an awareness that recent advances in electronic and information technologies can tremendously help improve the human's quality of life. As seen from the IEEE Transactions on Biomedical Circuits and Systems, one of the top journals in the field, publications on the designs of AFEs for biosignal appear in almost every issue of the journal since its first issue in 2007. Publications on AFEs also regularly appear in the IEEE Journal of Solid-State Circuits, the most prestigious journal in the field of integrated circuit design. These publications usually involve the designs of high-performance energy-efficient AFEs for recording different biosignals which include

neural signals (spikes and local field potentials or LFPs) [1-6], electroencephalogram (EEG) and electrocorticogram (ECoG) [7-9], electromyogram (EMG) [10, 11], and electrocardiogram (ECG) [12, 13]. Even though different biosignals exhibit somewhat different characteristics such as the frequency contents and the signal's amplitude, the specifications of AFEs for these biosignals have a lot in common. Thus, we should view the literature on AFEs for biosignal recordings as an integrated body of knowledge, instead of being specific to a particular type of signal. For instance, power and noise minimization techniques for neural signal AFEs can directly benefit the design of AFEs for cardiac signal monitoring.

Current AFEs are normally designed with a gain of 30-40 dB to mitigate the effects of noise from subsequent signal processing stages (filter, discrete-time amplifier, and analog-to-digital converter). Due to the limited supply voltage in advanced IC technologies, the AFEs need to reject the electrode offset while preserving the desired signal. In 2002, Harrison et al. [1] proposed a low-power neural amplifier for amplifying neural spikes and local field potentials. This work laid a foundation for the designs of fully-integrated biosignal amplifiers that followed. The AFE in [1] utilizes a capacitive feedback scheme with on-chip capacitors to realize a gain of 40 dB. The electrode offset is rejected with a capacitive coupling technique that uses very high-resistance MOS-Bipolar elements to create a high-pass corner at a frequency much lower than 1 Hz to preserve the desired signal. The techniques result in an AFE that is small enough to be implemented entirely on chip with good gain accuracy due to negative feedback. Subsequent works on neural-signal AFEs widely adopted the DC-offset rejection technique in [1] and made significant improvements in the AFE's energy efficiency [2, 5, 14]. However, this capacitive coupling scheme results in low input impedance which can affect the sensitivity of the recording due to signal attenuation at the input. Furthermore, there is a strong trend toward the use of dry electrodes instead of adhesive electrodes for patient's comfort [15-17]. Since dry electrodes exhibit much higher electrode impedances compared to adhesive electrodes, a high input-impedance AFE is needed to achieve a good sensitivity while using dry electrodes. As a result, AFEs for cardiac monitoring normally utilize different AC coupling techniques that allow the electrodes to be coupled directly to the gates of the AFE's input transistors. For example, the work in [12] uses an active feedback scheme that integrates the DC offset voltage at the output of the AFE and feed the correction current back to counteract the effect of the electrode offset; placing an integrator in the feedback path results in an AFE with a DC voltage gain

of zero. However, to keep the input-referred noise of the AFE low, the fed-back correction current is only limited to a small fraction of the bias current in the input transistors. Therefore, such active feedback scheme is only capable of rejecting the electrode offset voltage of approximately ± 50 mV. Though such offset cancellation scheme might work in most applications, it can fail in the cases when the input offset voltage is too large. To be highly robust across most recording situations, the AFEs should be able to reject rail-to-rail electrode offset. To accomplish this goal, the authors of [18, 19] used passive high-pass filters at the input. To avoid lowering the AFE's input impedance, they used positive feedback to inject the nulling current into the input of the AFE such that, effectively, the AFEs draw very small current from the signal's source. However, component mismatches in the input high-pass filters tend to reduce the common-mode rejection ratio (CMRR) of the AFE. To maintain a high CMRR, a large input capacitance is needed resulting in a very large chip area, which makes it very expensive for our research and development. For instance, the design in [19] uses an on-chip coupling capacitor of 1.2 nF, whose size alone is larger than the entire 1.5mm x 1.5 mm chip that we plan to fabricate. Thus, we need a different DC offset rejection scheme that allows the AFE to work robustly, while still being affordable.

Since the frequency content of ECG is at low frequency (0.1 Hz – 100 Hz) where MOS transistors exhibit high contents of flicker noise ($1/f$ noise). The $1/f$ noise should be removed to provide a clean recording. The most widely-used technique for removing $1/f$ noise in low-frequency biosignal AFEs is a modulation technique called chopper stabilization [20]. This technique has been applied extensively in neural field potential recordings [3, 21] and for other biosignals as well [8, 9, 12, 22]. However, the drawback of the chopping technique is the presence of the voltage ripple at the output of the AFE. This ripple occurs at the chopping frequency due to the up-modulated $1/f$ noise and the AFE's offset voltage. Thus, most chopping AFEs often require a scheme to minimize the output ripple. The authors of [23, 24] used a feedback technique similar to the electrode offset cancellation scheme discussed earlier. The feedback networks sense the voltage ripples at a point along the signal path and provide an appropriate nulling current to the input of the AFE; the high loop gain of this negative feedback loop results in the attenuation of the output ripple. This technique was shown to achieve the attenuation of the output ripple by as much as 60 dB. Other ripple minimization techniques include adding a notch filter at the ripple frequency in the forward path of the AFE [25], employing the binary search algorithm

to select the best combination of the input transistors that minimize the input-referred offset voltage [26], and using the binary search algorithm to adjust the optimum correction current that minimizes the ripple [18]. The aforementioned techniques have different advantages and drawbacks. The technique in [23, 24] uses feedback networks that operate continuously, thus it is able to continuously suppress the ripple but may consume extra power consumption due to the constant operation of the feedback networks. The binary search algorithms in [18, 26] adjust the AFE's intrinsic offset during the foreground calibration (before the input signal is applied), and is shut down once the calibration is completed. Since the calibration is only performed briefly at the start of the operation and then shut down, power consumption is lower. However, the need for calibration adds complexity in real applications; calibration must be done occasionally to compensate for drift in the offset voltage due to the change in temperature.

To make the entire cardiac monitoring device highly-portable while capable of recording weak ECG signals, the AFE must be very power efficient; it must exhibit low input-referred noise while consuming minimal amount of power. The dominant noise source of an AFE is usually the operational transconductance amplifier (OTA) used as a high-gain element. With the chopping technique, thermal noise, instead of $1/f$ noise, of the OTA dominates the overall input-referred noise of the AFE. As a result, the designs of the OTA usually focus on achieving the lowest input-referred thermal noise with the minimum amount of power consumption. The figure of merit (FOM) that is frequently used for comparing the power efficiency in the AFEs is the Noise Efficiency Factor (NEF), introduced by Steyaert et al. [27]. An ideal bipolar junction transistor (BJT) has an NEF of 1, while practical circuits with more number of devices that contribute noise have higher NEF's. The AFE in [1] was the first to achieve the NEF below 5, however, it requires a supply voltage of 5 volts to operate some transistors well into the saturation region. The PI's work, Wattanapanitch et al. [2], achieved a significant improvement in the NEF of 2.67, which is close to the theoretical limit of any AFE that uses a differential pair as an input stage. This work is currently the most cited work in the IEEE Transactions on Biomedical Circuits and Systems. Shortly thereafter, complementary input with current reuse technique has been introduced [28] which has lower theoretical limit on the NEF. However, due to its single-ended and open-loop nature of the design, CMRR, PSRR (power supply rejection ratio), and linearity of the AFE in [28] is too poor to be useful in real applications. Nevertheless, such idea gave rise to the complementary input differential

pair topology that combines excellent power efficiency of the current reuse technique and good CMRR and PSRR of the differential input pair topology. The works by [29, 30] used the complementary input technique as an input stage of the OTA, and used capacitive feedback as in [1, 2] for good gain accuracy and linearity. With such input pair technique, the AFEs were able to achieve an NEF less than 2. However, complementary input pair may place a limit on minimum power supply voltage to keep the input transistors in the saturation region. Also, the DC gain of the OTA with a complementary input pair is lower due to more parallel devices in the input stage. The decrease in the DC gain can mitigate the gain accuracy of the closed-loop amplifier. For instance, the work by [30] achieved an open loop gain of 70 dB, which is not enough to get a gain accuracy of 0.1% for an AFE with a closed-loop gain of 100. Thus, to achieve a good gain accuracy and an excellent NEF at a low supply voltage, a new OTA topology is needed.

Mains interference (50-Hz power-line noise) is a major concern for recording weak biosignals. Practical biosignal AFEs use differential-input topologies to achieve a high CMRR to reject mains interference by assuming that the interference couples almost equally to the two inputs of the AFEs. However, this is not always the case. Electrode impedance's imbalance can result in common-mode signals being converted into differential-mode signals, mitigating the usefulness of high-CMRR AFEs. Therefore, the works on cardiac AFEs strive to achieve high input impedances and high CMRR at 50 Hz; high input impedance ensures that imbalance in electrode impedances does not convert common-mode signals into differential-mode ones and high CMRR ensures that the common-mode interference is properly rejected. The works in [12, 18, 22] achieve a CMRR at 50 Hz on the order of 120 dB through fully-differential chopper topologies, and input impedance on the order of $G\Omega$. The CMRR at 50 Hz can be further improved by using the driven-right-leg circuits (DRL) [31, 32], which negatively feeds back the common-mode voltage to the patient's right leg through a third electrode. This negative feedback technique reduces the common-mode interference on the patient, thus improving the overall CMRR. However, in real situations, numerous coupling paths from the mains power to the electrodes on the body exist where each path exhibits different characteristics. Thus, we should assume that differential-mode signals due to mains interference are always present at the AFE's inputs and must be handled properly. A promising approach for cancelling the 50-Hz mains interference is to use feedback to add a nulling signal with the same frequency and magnitude as the interference, but with a 180 degree phase shift, to

the input [33, 34]. This approach effectively creates a notch in the AFE's transfer function at the frequency of the interference. However, current implementations of this approach require power-hungry circuits such as a phase-lock loop and analog multipliers [33] or require an off-chip digital processor to implement full-blown digital filters [34], thus making them unsuitable for ultra-low-power systems. Therefore, a new approach that can efficiently create a notch transfer function at the interference's frequency with minimum additional circuitry and power consumption is needed.

Motion artifact (MAF) in the recorded data due to patient's movements is a major concern for wearable cardiac monitoring devices. The MAF can be very large relative to the desired signal and can saturate the output of the AFE, causing information loss. To prevent such saturation, the AFE should be able to reject the MAF right at the front end. It has been shown that MAF correlates well with variations in the electrode impedances which are caused by the patient's movements. Thus, adaptive filters can be used to map the variation in electrode impedances to the MAF and subtract it out such that the recorded ECG is virtually free of MAF. There are a number of publications that use adaptive filters with simple least mean square algorithms (LMS) to perform MAF removals [35, 36]. However, these algorithms were implemented in software at the backend which require that the AFE have a wide dynamic range. Such AFEs are challenging to design and consume more power, thus are not suitable for our intended low-voltage low-power design. To solve this problem, Kim et al. [19, 37] proposed a topology that can remove the MAF right at the front end. The topology performs adaptive filtering on the digitized data using a low power microcontroller (MSP430 from Texas Instrument). The microcontroller feeds the MAF's information back to the input of the AFE through an on-chip digital-to-analog converter (DAC) to subtract it from the original signal. Due to the ADC and DAC embedded within the MAF removal feedback loop, there is a long latency in the loop that creates signal distortions. Furthermore, the need for a high-resolution DAC (12 bits in [19]) adds significant area and power overheads to the overall system. A better solution might be to implement a simple adaptive filter in analog domain, which eliminates the needs for ADC and DAC in the feedback loop. This method can significantly reduce latency and conserve power. The PI's previous work on an analog adaptive filter [38-40] used a Gm-C topology to implement a simple LMS algorithm for decoding neural signal in brain-machine interface applications. The filter consumes sub-microwatt of power and will be researched further for use in this proposed project.