Smart sensor interface in biomedical monitoring systems

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Introduction

Current trends in biomedical electronics target the development of portable battery-operated wearable devices for remote monitoring and patient care in a ubiquitous healthcare environment [1, 2]. Some typical applications for ubiquitous healthcare are health monitors, which include biopotential monitors, monitoring of epileptic conditions, breath monitors, cough monitors, polysomnography, activity monitors, etc., as well as control of medication delivery, assisted rehabilitation, etc. [3]

Remote patient monitoring envisions the reduction of hospital cramming and enables patients to self-manage their health conditions in a continuous fashion [4]. One step forward, continuous monitoring of the health conditions enables an early detection of certain diseases, which might remain otherwise undiagnosed in case of only rare visits to the hospital as is with hospital-centered healthcare.

Each sensor, with its corresponding interface, operates towards the acquisition of biomedical signals. Taking into account the different nature of the signals being monitored, there are a variety of sensors and sensor interfaces to consider for the biomedical monitoring system. A sensor interface front-end typically performs some analog processing operations for signal conditioning, followed by analog to digital converter (ADC) and some further signal-specific digital processing [5]. Accordingly, the processing parameters must be adapted in order to have the processing chain suited to the nature of the biomedical signal.

This work handles the analog processing section of a sensor interface. Analog processing accounts mainly for gain and filtering. The signals sensed by the sensors resemble very small amplitudes, thus an instrumentation amplifier (IA) with very large gain is required to bring the signal levels into the dynamic range of the latter analog processing sections. The next processing operations consist of some filtering, e.g. highpass, lowpass and notch filters. Indeed, the IA will also amplify noise and interferers besides the biomedical signal of interest. These should be filtered out resulting in an improvement of the biomedical signal’s signal to noise ratio (SNR), compared to the signal prior to signal conditioning. This relaxes the design specifications for the following ADC.

Another issue to be taken into account regards signal integrity. In a ubiquitous healthcare environment, the biomedical signal recorded by the wearable monitoring device is affected by a series of artifacts generated by patient movement, improper electrode attachment, loose conductors, etc. [6, 7]. These manifest in an additive manner and distort, or even corrupt, the features of the biomedical signal. For this purpose, artifact suppression should be performed as early as possible in the signal processing chain. Artifact suppression is commonly performed in the digital domain. However, having the artifacts reduced via analog filtering prior to A/D conversion will contribute to relaxing the design specifications for the ADC.

As illustrated, the processing parameters of the sensor interface analog front-end must be properly set in order to have the sensor interface adapted to the specifics of the biomedical signal. One step forward, having the processing parameters adapted to the nature of the biomedical signal in an automatic fashion implements some form of intelligent adaptivity into the sensor interface. This gives the definition of a smart sensor interface as the front-end
circuitry with some embedded intelligence, performing signal conditioning and self-calibration [8] in response to the particularities of the biomedical signal.

**Materials and Methods**

A generic front-end of a smart sensor interface is illustrated in figure 1. The analog signal processing chain consists of an instrumentation amplifier which delivers a large gain to the biomedical signal, followed by a notch filter for mains frequency suppressing, a highpass filter to suppress the DC component and a lowpass filter to separate the in-band signal from out-of-band noise and interferes. Additionally, an amplifier in the signal pass performs amplification/compression, in either linear or non-linear fashion, of the biomedical signal.

![Figure 1. Generic front-end of a smart sensor interface](image)

Portability and battery operation of the biomedical monitoring equipment impose some critical features to the analog circuitry expressed in the need for increased processing power, long battery lifetime, as well as small size specifications [3]. A lightweight and small device is obtainable through miniaturization in the context of More Moore and More than Moore. Processing capabilities and small power consumption on the other hand account for severe design constraints, expressed in terms of low-voltage supply, low-power consumption, linearity, low noise, etc. [5]

A Gm-C implementation was chosen to implement the processing sections of the analog front-end. The basic building block of Gm-C circuits is the operational transconductance amplifier (OTA). The OTA presented in this work is illustrated in figure 2 and is described as follows [9].

![Figure 2. Transistor-level implementation of the OTA](image)

The OTA was designed for a 1.2 asymmetric supply voltage and subthreshold biasing archives a power consumption in the micro-power range, i.e. 1.32μW [10]. The OTA illustrated in figure 2 is resembles a cascade of two amplification stages [11]. The first stage is implemented with a linearized transconductor core consisting of differential input pair M1-M2 linearized with bulk-input, source degeneration MD1-MD2, gate degeneration and bump linearization MB1-MB2. The linearized transconductor core is biased with DC current source Ibias1 and loaded with diode-connected transistors M3-M4 [5]. The second stage is
implemented with differential input pair M5-M6 biased with DC current source $I_{bias2}$ and loaded with active resistances $R_L$. Transistors M3, M4, M5, M6 implement a translinear loop, which enable the expression of the OTA transconductance as the ratio of the bias currents [11]:

$$G_m = G_{m,in} \cdot \frac{I_{bias2}}{I_{bias1}}$$  \hspace{1cm} (1)

which makes transconductance tuning both simple and straightforward.

Programmability of the OTA transconductance via the bias current ratio is extrapolated to programming the processing parameters of various analog processing sections. This is illustrated in figure 3 for a variable gain amplifier (VGA), envelope detector, an analog biquad and an allpass filter [9,10].

The parameter adaptation block in figure 1 implements an intelligent algorithm to control the analog processing parameters. Accordingly, the parameters of the analog processing chain are adapted in a feed-forward control loop in order to meet the specifications of the signal being monitored.

There are two approaches for processing parameter adaptation. The first method requires a reference signal, and the parameter adaptation procedure aims to adapt the analog processing parameters in order to fit the monitored signal features to those of the reference signal. For this purpose, the features of the monitored biomedical signal are extracted in terms of means of power spectral density, spectrogram analysis and average power [7]. Next, the parameter adaptation algorithm operates towards minimizing an objective function defined as

Figure 3 Gm-C analog section: (a) variable gain amplifier, (b) envelope detector, (c) biquad and (d) allpass filter.
\[ \varepsilon = |\text{features}^{\text{ECG}} - \text{features}^{\text{ref}}| \]

which accounts for the deviation from the reference signal features \[12\]. As a result, the SNR of the monitored signal is increased. The second method for parameter adaptation analyzes the monitored signal solely and aims to increase the SNR, which defines the objective function in this case.

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