An Adaptive ENG Amplifier for FES Applications

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Abstract

Electroneurogram (ENG) recording from tripolar cuff electrodes is degraded by interference signals mostly generated from muscles nearby. Interference reduction may be achieved by suitably designed amplifiers such as the true-tripole and quasi-tripole configurations. However, in practice the performance of these amplifiers is degraded by the presence of cuff imbalance. This paper describes an alternative amplifier configuration, termed the adaptive tripole, developed to automatically compensate for cuff imbalance and thus improve the quality of the recorded ENG. The three amplifier configurations are compared in terms of their output signal-to-interference ratio in the presence of cuff imbalance. In-vitro experiments, using a saline bath setup, indicate that the adaptive tripole performs much better and even better when it is integrated.

1 Introduction

Tripolar cuff electrodes are appropriate for recording electroneurogram (ENG) signals chronically after implantation [1], as they can be made biocompatible and they increase the signal amplitude. Because the amplitude of the neural signal inside the cuff is usually only a few µV, it may easily be overwhelmed by interference or noise. A major source of interference is the myoelectric activity of nearby muscles (the electromyogram or EMG) [2]. The EMG amplitude is approximately three orders of magnitude larger than the µV ENG and their spectra overlap.

Ideally, using appropriate amplifier configurations for reducing EMG, naturally occurring ENG can be recorded and used as control input in FES devices [3] for correcting gait, providing grasp or controlling the urinary bladder after spinal cord injury [3]-[5]. An example of a possible block diagram of an implant using ENG recording for stimulation control is shown in figure 1.

Figure 1: Example of a Bladder Stimulation Implant

An RF link can be used to allow external control and monitoring and for providing the power supply using batteries in the external unit.

The recording circuit has to provide a clear ENG signal to be used for triggering the stimulation. However the ability of the conventional amplifier configurations [6] to reduce EMG is reduced as the tripolar cuff departs from its ideal performance. Cuff imbalance is introduced between the two halves of the cuff, due to asymmetry and tissue growth [7]. As a result the SIR (signal to interference ratio) at the output of configurations like the quasi-tripole (QT) [2] and the true tripole (TT) [6] is usually much less even than 1:1 and ENG detection is difficult when active muscles are present in the surrounding area of the recording cuff.

The recently introduced “adaptive tripole” (AT) was developed to balance the cuff and remove EMG interference [8]. It is based on the TT structure with the addition of an adaptive gain stage and a control system for allowing it to automatically adapt to cuff imbalance changes. This paper describes saline - bath experiments performed on discrete component versions of the AT and the other systems for comparison. Moreover an IC version of the AT is tested for demonstrating the advantages of using ASICs in implantable devices.
2 Methods

2.1 System description

The block diagram of the AT is shown in Figure 2 and it consists of low noise differential pre-amplifiers (gains $A_1$ and $A_2$), followed by amplifiers with variable gains $G_1$ and $G_2$ controlled by the differential feedback signals $a_1$ and $a_2$. The AT operates by first rectifying the outputs of $G_1$ and $G_2$ and applying them to the comparator whose output is applied to a long time-constant integrator, which generates the differential feedback signals $a_1$ and $a_2$. The variable gain amplifiers counterbalance the cuff error (imbalance) by equalizing the amplitudes of the two pre-amplified input signals $V_i_1$ and $V_i_2$. Once the integrator outputs have settled, the ENG can be extracted at the output ($V_{out}$) of the summing amplifier $G_{out}$.

In the integrated version of the system, fabricated in 0.8µm BiCMOS technology, the variable gain amplifiers are replaced by variable transconductance OTAs (operational transc. amplifiers) and the rectifiers, as well as the integrator and the feedback stage operate in current-mode, making operations like signal addition much easier.

2.2 Definition of cuff imbalance

If the cuff in Figure 2 was balanced, then $V_{i1} = V_{i2} = 50\% V_{EMG}$. However, in the presence of cuff imbalance (denoted here by $X_{imb}$), $V_{i1} = 0.5 (1 + X_{imb}) V_{EMG}$ and $V_{i2} = 0.5 (1 - X_{imb}) V_{EMG}$, where $|X_{imb}| < 100\%$. Thus,

$$X_{imb} = \frac{V_{i1} - V_{i2}}{V_{EMG}} \times 100\% \quad (1)$$

In practice, $X_{imb}$ can be calculated by the preamplifier outputs $V_{A1}$ and $V_{A2}$:

$$X_{imb} = \frac{|V_{A1} - V_{A2}|}{2 \left| V_{A1} + V_{A2} \right|} \times 100\% \quad (2)$$

2.3 Saline bath setup

The system has already been tested using transformer-coupled inputs (EMG + ENG). In this experiment, more realistic tests were performed using a saline-bath (Figure 3) containing a hand-made tripolar recording cuff. The EMG field was generated by a bipolar cuff with insulation between the electrodes, an easy way to form a dipole. The signal was generated by noise band-limited between 50Hz and 3kHz, with maximum amplitude of over 1mV. The ENG signal was a 1kHz sinusoid with amplitude of 3µV and it was injected by a transformer in series with the middle electrode. Cuff asymmetries were present and different recording cuffs with different asymmetries were used to test the performance of the systems for different values of imbalance.

![Figure 2: The AT, system-level block diagram. Gains of $G_1$ and $G_2$ are controlled by the feedback to minimise cuff imbalance. Interference is eliminated at the output stage $G_{out}$ and $V_{out}$ is the extracted ENG.](image)

![Figure 3: Top view of the saline bath](image)
3 Results

Figure 4: Switching from the TT to the AT ((a) at t=10s) reduces interference. The signal is detectable at the AT output (c).

Figure 4a illustrates the output of the amplifier circuit over 20s, with the configuration switching from the TT to the discrete-component AT at half time (10s). The random EMG component at the system output is reduced. An inspection of the outputs of the two systems over 40ms indicates that the ENG, which is the high frequency component at the output, can be detected when the system operates as the AT (Figure 4c).

Figure 5: SIR comparison between the AT and the two other systems.

A comparison of the signal-to-interference ratios of the three systems is shown in figure 5. The solid lines represent the IC version of the AT, while the dotted lines are used for the data obtained from the discrete-component version. The QT SIR was on average 1.18 times greater than the TT and the discrete AT was on average 15.4 times greater than the TT, while the corresponding figure for the integrated version was 86.2, the latter resulting to 5.6 greater average SIR than its discrete-component version.

4 Discussion and Conclusions

The use of a proper tripolar amplifier configuration can lead to fully implantable ENG controlled FES systems. The AT offers much higher SIR than its fixed-gain counterparts, as it dynamically compensates for cuff imbalance. The results presented illustrate the corrective action of the AT relative to the TT and the QT. In figure the interference reduction when switching from the TT to the AT is evident and the ENG can be detected only when the AT is used. Using the saline bath setup allows for a more realistic test compared to transformer-coupled testing, where the cuff is represented by purely resistive components.

The AT has been implemented both in discrete and in integrated form using analogue integrated circuit techniques that result in a fully implantable ENG amplifier. The integrated version performs 5.6 times better than the discrete version as a result of better matching between the components and it consumes less power and occupies less space.

References


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