ADAPTIVE INTERFERENCE REDUCTION IN NERVE CUFF ELECTRODE RECORDINGS

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ABSTRACT

Neural signals (ENG) recorded from insulating cuffs fitted with electrodes and placed around nerve bundles may replace artificial sensors in providing feedback signals in *functional electrical stimulation* (FES) applications. Typical applications include correction of foot-drop, hand grasp in tetraplegic subjects and bladder voiding in certain types of incontinence. Unfortunately, the ENG signal recorded using this method is on the order of a few µV whereas interfering signals can have amplitudes of many mV. Probably the main source of such interference is the Electromyogram (EMG) which is generated by excited muscles near the cuff. A method is presented to reduce the interference pickup in nerve cuff recordings. The gains of differential amplifiers connected to true-tripole nerve cuff recording arrangement are tuned adaptively to null the residual EMG. Simulation results show that extraction of the neural signal is possible using this method without the need for additional high order filtering.

I. INTRODUCTION

Neural signals (ENG) recorded from insulating cuffs fitted with electrodes and placed around nerve bundles are replacing artificial sensors in providing feedback signals in *functional electrical stimulation* (FES) applications. Typical applications include correction of foot-drop [1] and hand grasp in tetraplegic patients [2]. Unfortunately, the ENG signal recorded using this method is on the order of a few µV whereas interfering signals can have amplitudes of many mV. The main source of interference is the electromyographic (EMG) potential generated by excited muscles near the cuff. Various methods have been suggested to overcome this difficulty, mostly based on the use of multiple electrode structures within the cuff [3,4]. Fig. 1 shows a cylindrical cuff fitted to a nerve bundle. In this example, the insulating cuff of length L and internal diameter D is shown fitted with three equally spaced circular electrodes. L is typically 2-3cm while D is typically about 1mm depending on the diameter of the nerve bundle. To a first approximation, the nerve bundle is an insulator, while the space between the nerve bundle and the cuff is filled with a conducting fluid.

The ENG signal results from the action potentials, propagating along the nerve fibres contained in the bundle, which cause small currents to flow through the fibre membranes to the extrafascicular medium. Within an insulating cuff, a thin sheet of conducting fluid exists between the cuff and the nerve bundle causing the local impedance to be much higher than outside the cuff. As a result the membrane currents just referred to give rise to small potentials which are measurable between the electrodes in the cuff. In addition, the cuff length must be equal to the wavelength of the transmembrane action potential, which is about 3cm for a 10µm fibre diameter and the outer electrodes must be placed at the ends of the cuff. Examination of the power spectral density (PSD) of nerve cuff recorded signals show the ENG signal spectrum lies mainly between 1 and 2kHz.

In contrast, the interfering EMG originates in excited muscles outside the cuff during electrical stimulation. The EMG has a spectral peak between 100 and 200Hz, with some energy still present up to 1kHz. There is no appreciable phase variation of the EMG signal over a region with the dimension of a typical recording cuff. Given this important fact, methods to reduce the pickup of EMG and other sources of interference have centred on the linearisation [5] of the internal field generated by external sources, i.e., EMG. Fig. 2 shows the potential variation along the whole length of the cuff, where V_1 and V_2 are the potentials at the ends of the cuff.

The interfering voltages recorded from a symmetrical tripolar electrode structure are equal and opposite and can be cancelled by a suitably designed differential amplifier arrangement. One such arrangement, called a *true tripole,* is shown in Fig. 3 [3].

In this arrangement, the dashed box represents the cuff, while the recording system consists of three amplifiers with gains G_1, G_2 and unity, respectively. Z_{t1} and Z_{t2} represent the tissue impedances inside the cuff and Z_{t0} is the tissue impedance outside the cuff while Z_{e1} Z_{e2} and Z_{e3} are the electrode-tissue contact impedances. However the various impedances are subject to significant variation with time for two main reasons:

1) changes in the nerve cuff impedances attributed to tissue re-growth inside the cuff produced after a period of implantation [6].

2) manufacturing tolerances.

If we assume that the input differential amplifiers have zero input currents, the output of the system in response to an external interfering signal such as the EMG is given by:

$$V_{EMG} = V_{\text{int}}\left[\frac{G_1 Z_{t1} - G_2 Z_{t2}}{Z_{t1} + Z_{t2}}\right] \qquad \ldots(1)$$

where V_{int} is the interfering signal appearing between the ends of the cuff. Note that V_{EMG} can be made zero by adjusting G_1 and G_2 irrespective of the cause of the tissue impedance mismatch. However, examination of eqn. (1) shows that the interference cancellation is very sensitive to the parameter values, emphasizing the significance of the imperfections referred to above. Practically speaking, in order to use the true-tripole arrangement to best advantage, some form of adaptive adjustment of G_1 and G_2 is necessary.

In this paper, we present an adaptive control system, which can be implemented in low power BiCMOS technology. This system will dynamically cancel errors in the true tripolar recording system by automatically adjusting *G1* and *G2* to force the numerator of equation (1) to zero, irrespective of their origin. The development of the circuits to be ulitimitely implanted are at an early stage and we present simulations based on MATLAB/SIMULINK. In addition, we describe some preliminary work on the design of a low noise preamplifier, which will form the interface between the nerve cuffs and the control system described above.

II. THEORY

A simplified block diagram of the proposed control system is shown in Fig. 4. It consists of a true-tripole recording structure in which the gains of the input differential amplifiers, *G1* and *G2* have been made variable, controllable by the differential feedback voltages *Vc+* and *Vc-*. The proposed method realises a modified form of the Hérault-Jutten (HJ) algorithm for *blind estimation of sources* [8,9,10]. Since the spectra of the two signals (ENG and EMG) are separated in frequency, the integrator time constant can be chosen in advance. No training procedure is required, thus simplifying the required hardware realization.

The system operates by obtaining first the moduli of the outputs of the differential amplifiers *G1* and *G2* and comparing them to determine which is the largest. The Compare and Select (C&S) block provides a voltage output of value $\pm Ve$, the sign depending on which input is the larger. So, for example, if the output from *G1* is larger than *G2*, the output from the C&S block is +Ve. This is integrated by the differential output integrator block and the control voltages *Vc+* and *Vc-* are incremented by equal and opposite amounts to compensate for the error. Simulations have shown that the system is capable of correcting large static errors in the ampltudes of the inputs within 10 to 15 ms. In addition, although a long time constant integrator is required the time constant itself is not critical and so an integrated solution is feasible.

The system as described above is intended to compensate for static and/or slowly time varying changes in the characteristics of the source. However, the ability of the system to eliminate interference such as EMG depends on the linearisation property described above. In practice, since the aspect ratio of the cuff cannot be made infinitely large, linearistion will not be perfect, resulting in residual interference artifacts appearing at the output. In the past, these have been removed by frequency domain methods, specifically by band-pass filtering. In the absence of any correction of manufacturing tolerances, etc., this can require high levels of filtering: the use of a 91-pole digital filter having been described [7]. Such solutions are clearly impractical for integrated, implantable systems.

In principle, since residual artifacts appear at the input to the system as amplitude errors *only*, it is possible to use the control system to cancel these artifacts dynamically, eliminating the need for frequency domain filtering. It should in theory only be necessary to reduce the integrator time constant appropriately. The implementation of such a system, which does not rely on linearisation and, therefore, on the implantation of long cuffs, should allow the use of shorter, more practical cuffs.

III. REALISATION

The block diagram of Fig 4 is realised in the modified form shown in Fig 5. The main differences between figures 4 and 5 is (a) that the first-rank amplifiers are split into two cascaded sections and (b) that the compare and select block has been replaced by a finite gain amplifier with differential inputs.

The first section (preamplifier) of each first-rank amplifier has a fixed gain of about 100 (40dB) and is optimized for low noise operation. The second stage has a nominal gain of 5, variable in the range 1 to 10 (the second-rank amplifier has a fixed gain of about 200, so that the ENG output signal has an amplitude on the order of 100mV). A BiCMOS preamplifier has been designed using the CADENCE design tools. Its basic specification is shown in Table 1.

It was found that the use of a high-gain comparator in the compare and select block in Fig 4 led to instability about the equilibrium point, i.e., when the gains had been correctly adjusted and the system was 'idling'. Reducing the loop gain removed the instability with negligible loss in accuracy. The modulus circuit can be conveniently implemented by means of a precision full-wave rectifier.

IV. SIMULATION RESULTS

The behavior of the control system shown in the block diagram of Fig 4 was investigated with several different input signals using MATLAB/SIMULINK. Fig. 6A shows the output of the system when $x_1(t)$ (ENG) and $x_2(t)$ (EMG) are a sine wave (1μV, 1600Hz) and a square wave (1mV, 200Hz), respectively. The system was able to extract $x_1(t)$ in about 15ms when the impedance mismatch was 5% and the integrator time constant was 0.5s.

Fig. 6B shows the output of the control system when $x_1(t)$ and $x_2(t)$ are a sine wave (1μV, 1600Hz) and random noise (1mV), respectively. The output converged to a signal proportional to $x_1(t)$ within 15ms for an impedance mismatch of 5%.

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Table 1
Preamplifier Specification

Voltage gain	100
Bandwidth	15KHz
Common mode rejection ratio	100dB
Equivalent input voltage noise	$2.5nV/\sqrt{Hz}$
Equivalent input current noise	$2.8pA/\sqrt{Hz}$
Power supplies	±2.5V
Power consumption	1.2mW

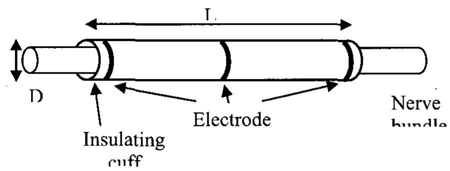

Fig 1: An insulating cuff and tripolar electrode assembly fitted to a nerve bundle.

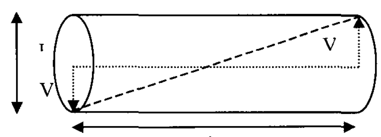

Fig 2: Idealised potential variation inside the cuff

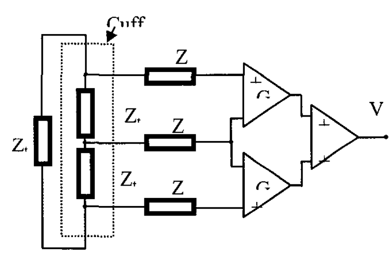

Fig 3: *True-tripole* recording system and amplifier arrangement

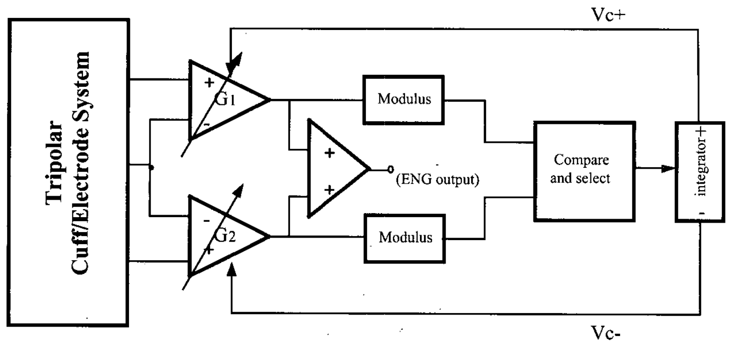

Fig 4: Elements of an adaptive controlsystem for a *true tripole*

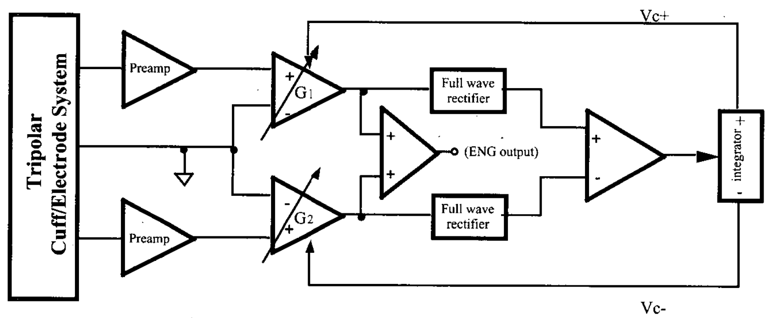

Fig 5: Improved version of Fig 4

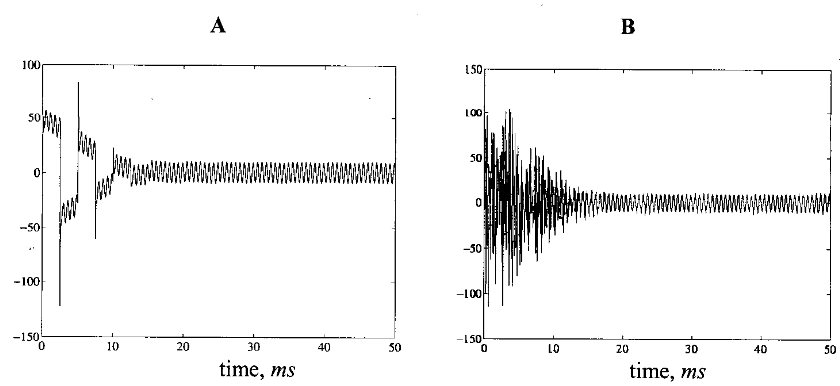

Fig 6: Simulated perormance of the control system.
A: $x_1(t)$ *sine wave* and $x_2(t)$ square wave. B: $x_1(t)$ *sine wave* and $x_2(t)$ random noise.