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Comparison of Three ENG Tripolar Cuff Recording Configurations

I. F. Triantis[1], A. Demosthenous[1] and N. Donaldson[2]
[1] Department of Electronic and Electrical Engineering, University College London, UK
[2] Department of Medical Physics and Bioengineering, University College London, UK

Abstract- **Cuff electrodes are a safe and reliable technique for long-term recording of the electroneurogram (ENG) from peripheral nerves. However, the usefulness of the recorded ENG depends on the amount of electromyogram (EMG) and stimulus artifact present. Interference cancellation may be offered by tripolar cuff recording configurations, such as the quasi-tripole (QT) and the true-tripole (TT), but in practice their performance is severely degraded by cuff imbalance. The adaptive-tripole (AT) has been developed to compensate for possible cuff imbalance and hence minimize the interference pick-up on the recorded ENG. In this paper, the signal-to-interference ratios of the three tripolar cuff recording configurations are compared in-vivo in the presence of cuff imbalance. The results of the experiments, which were conducted using stimulation-induced signals, indicate that even though two different imbalances were present, the AT outperforms the other two configurations in the majority of cases.**

Keywords - **Cuff imbalance, ENG recording, EMG, tripolar electrodes, adaptive-tripole.**

I. INTRODUCTION

Neuroprostheses may use naturally occurring neural signals (ENG) for improving functional electrical stimulation (FES). Typical FES applications include correction of foot-drop, hand grasp in tetraplegic patients, and bladder voiding [1]-[3]. In these applications insulating cuffs incorporating metal electrodes are used to record ENG activity. Nerve cuff electrodes are suitable for chronic implantation [4] but the usefulness of the ENG recorded from them depends on the amount of electromyogram (EMG) interference and stimulus artifact present. To reduce interference the cuff electrodes are made tripolar and symmetrical where three equally spaced electrodes are placed along the inside surface of the cuff.

A tripolar cuff may be used with the quasi-tripole (QT) [5] or the true-tripole (TT) amplifier configuration [6]. In both structures, EMG cancellation is achieved by exploiting the *linearization* effect of the insulating cuff on the internal fields [7]. In theory, by suitably arranging the electrodes in both amplifier configurations, the EMG is cancelled and only the ENG is recorded. In the QT the two outer cuff electrodes are connected to one input of a differential amplifier and the middle electrode is connected to the second input. This ideally forces the EMG potential at the middle electrode to be equal to the potential at the two outer electrodes so that they cancel at the output of the amplifier. In the TT the two outer cuff electrodes are connected to two separate differential amplifiers and the middle electrode to the other input of both amplifiers. The outputs from the two amplifiers are then summed in a third amplifier where ideally the two EMG potentials are of opposite polarity and will cancel out.

One benefit of the TT is that the ENG amplitude recorded is about twice that in the QT. On the other hand, the TT is more severely affected by cuff imbalance than the QT. The cuff imbalance might be a result of manufacturing tolerances in the placement of the electrodes in the cuff *or* tissue growth in the cuff after implantation [7].

In the past, attempts have been made to separate the ENG and EMG signals by filtering, since the peaks of their power spectral densities differ by about an order of magnitude (although the spectra overlap considerably). This frequency-domain approach requires very high-order digital filters [8], a solution which is not easily implantable [8]. A fully implantable low power ENG amplifier may be offered by the adaptive-tripole (AT) [7], [9], which has been developed to compensate for large cuff imbalances and to minimize EMG breakthrough in ENG recordings without the need for complex filtering.

This paper presents a comparison of the QT, TT and AT tripolar cuff recording configurations in terms of *signal-to-interference* (*S/I*) ratio. The comparison is based on in-vivo measurements using stimulation-induced signals for different values of cuff imbalance. The remainder of the paper is organized as follows: The operating principle of the AT configuration is described in Section II. Experimental methods and measured results are presented in Sections III and IV, respectively; this is followed by a discussion of the experimental findings in Section IV. Finally conclusions are drawn in Section V.

II. THEORY

The block diagram of the AT system is shown in Fig. 1. It consists of a TT structure with 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 obtaining the *moduli* of the outputs of G_1 and G_2 and applying them to the differential amplifier G_3. The resulting difference signal is applied to a long time-constant integrator, which generates the 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_{i1} and V_{i2}. Once the integrator outputs have settled, the ENG can be extracted at the output of the summing amplifier G_o (node marked V_{AT}), since the EMG will cancel at that node.

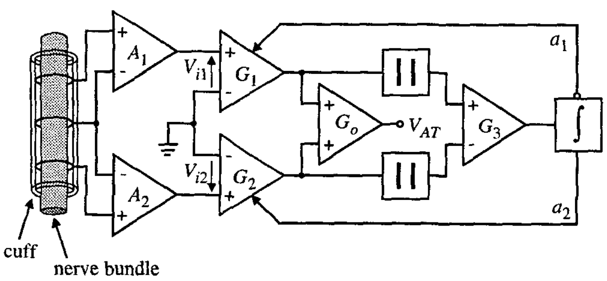

Fig. 1. The AT cuff recording system.

In this paper, cuff imbalance, X_{imb}, is defined as follows: Assume that the inputs V_{i1} and V_{i2} in Fig. 1 are solely interference and $|V_{int}|$ = $|V_{i1}|$ + $|V_{i2}|$, where V_{int} is the interference across the end electrodes of the tripolar cuff. Ideally:

$$|V_{i1}| = |V_{i2}| = 50\% \, |V_{int}| \tag{1}$$

However, in practice:

$$|V_{i1}| = (50\% \pm X_{imb}) \, |V_{int}| \tag{2}$$

with X_{imb} having a theoretical maximum value of 50%.

III. Methodology

A. Implementation of the Recording System

A discrete component PCB recording system was realized that allowed manual switching between QT, TT and AT amplifier configurations. The power supply rails were ±8V and these were extensively decoupled, and the PCB was placed in an aluminum box to minimize electrical noise. All components refer to Fig. 1. The preamplifiers were realized using AC coupled AMP01 differential amplifiers with a bandwidth of 25kHz and a gain of 400. The variable gain amplifiers were realized using AD633 analogue multipliers with a mean gain of 2. Each full-wave rectifier was built with an amplifier (OPA277) and two diodes (1N4148) and the amplifier G_3 was realized using a difference amplifier (OPA277) with a gain of 200. A first-order active integrator with a time-constant of 1s provided the differential feedback signals a_1 and a_2 in Fig 1. The choice of the integrator time-constant of 1s ensured that the output harmonic distortion on the recorded ENG was kept to a minimum. Finally, the overall ENG path gain was set to 16000 by appropriately choosing the gain of the summing amplifier G_o. The system was designed to compensate for X_{imb} values up to 40%.

B. Experimental Setup

Eight sets of acute experiments [10] were performed in six New Zealand White rabbits with weights of 3 to 3.5kg. The rabbits were anaesthetized with "rompun cocktail", first dose

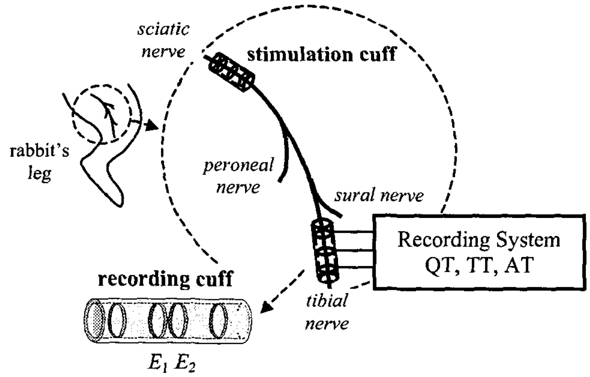

Fig. 2. Experimental setup. Illustration of the nerves involved and position of the recording and stimulation cuffs.

5ml and subsequent hourly doses of 2ml. A split type cuff [4] with inner diameter of 2mm and length of 22mm was used with four platinum foil ring electrodes (see Fig. 2). The two middle electrodes (denoted E_1 and E_2) were placed each side of the center of the cuff, 2mm apart and the remaining two electrodes were placed at 2mm from the cuff edges. The width of each electrode ring was 2mm. The middle electrode of the tripole cuff could be selected between E_1, E_2 or E_1E_2 (i.e., shunted together to represent a more balanced condition). Varying the middle electrode allowed some control over the imbalance.

A tripolar stimulating cuff with inner diameter of 2mm and length of 12mm was used to apply *bipolar* stimulation, with current pulses of amplitude 1–1.5mA, pulse width of 200μs and period of 0.2s. The stimulating cuff was placed around the sciatic nerve and the recording cuff around the tibial nerve approximately 2cm apart from each other (see Fig. 2). Measurements were taken for E_1, E_2 and E_1E_2, using each of the three tripolar cuff recording configurations. The results were captured on a TEAC RD-145T DAT recorder with a sampling frequency of 20kHz.

IV. Results

The interference signals were stimulus artifact and *M*-wave. They possessed *different* imbalances as shown in Fig. 3 and this was expected to cause some difficulty to the variable-gain selection of the AT system. The interference signal of interest was the *M*-wave, because the AT was not developed for operation with simultaneous stimulation. The compound action potential (CAP) was the useful ENG extracted. The peak S/I ratios (i.e., CAP peak over *M*-wave peak) of each tripolar cuff recording configuration (all eight experiments) for each middle electrode possibility (E_1, E_1E_2, E_2) are shown in Fig. 4.

As an example of the time-domain signals recorded, Fig. 5 shows the responses of the three recording configurations (experiment number 8, E_1E_2, E_2) scaled to have the same CAP amplitude (2^{nd} CAP peak) for easier visual comparison.

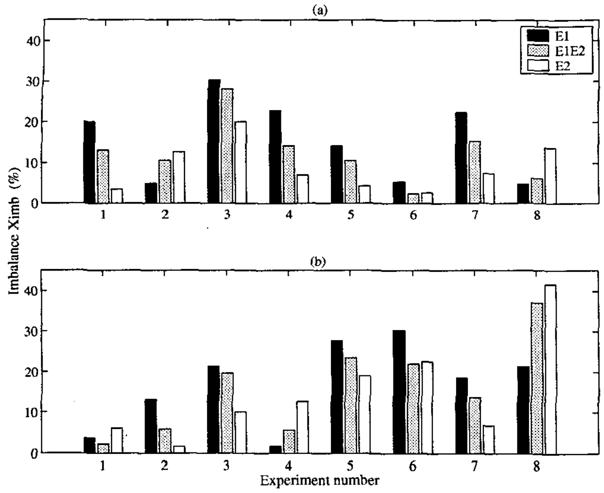

Fig. 3. X_{imb} variation for change between E_1–E_1E_2–E_2 (all experiments); a) For M-wave, b) for stimulus artifact.

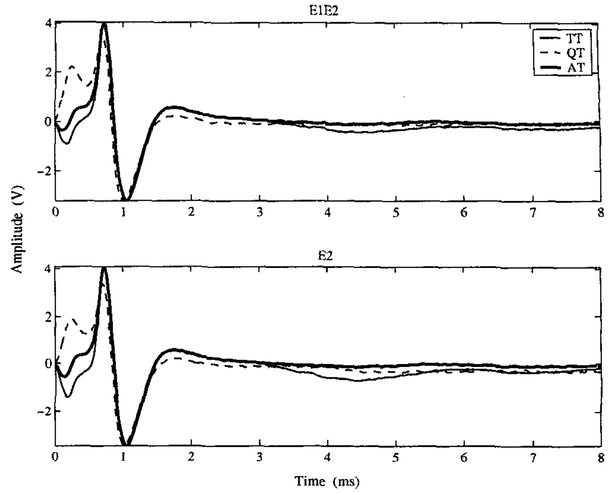

Fig. 5. Output responses of the three tripolar recording configurations (experiment number 8, E_1E_2, E_2).

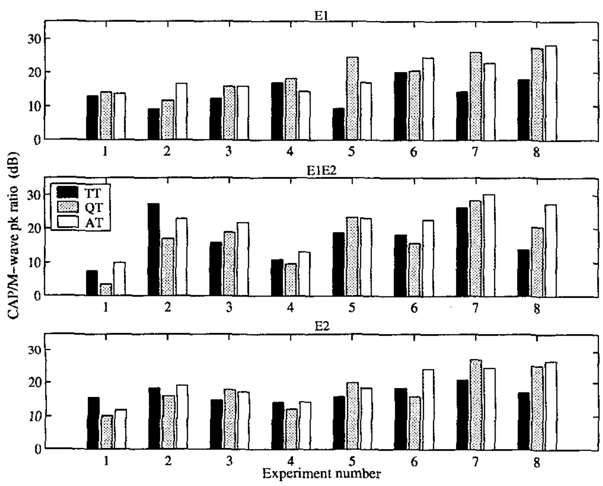

Fig. 4. CAP to M-wave ratios of the TT, QT and AT (all experiments) for change between E_1–E_1E_2–E_2.

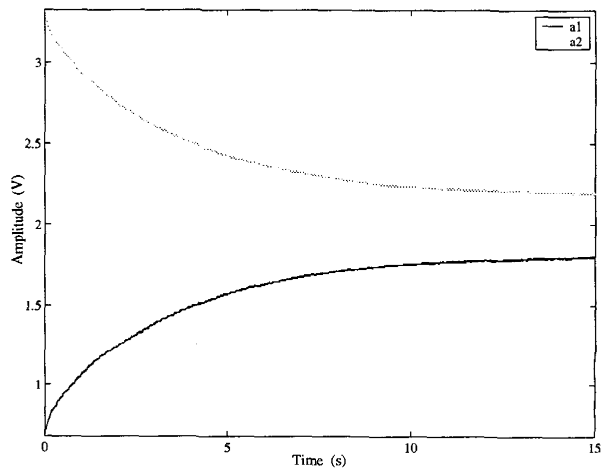

Fig. 6. Response of the differential feedback signals of the AT (experiment number 8, E_1E_2).

In both cases, the AT has the smallest stimulus artifact peak and the smallest M-wave peak (measured between 3 to 5ms).

Figure 6 shows a typical time-domain response for the feedback signal a_1 and a_2 of the AT (see Fig. 1). After an initial adaptation phase of 10s the feedback signals settle to the values required to compensate for the particular cuff imbalance (in this case X_{imb} = 5%). It should be noted that the waveforms in Fig. 5 were recorded after the feedback signals of the AT had settled to their near optimum values.

V. Discussion

As shown in Fig. 4, the TT had the greatest peak S/I ratio in only two out of twenty-four cases and it performed better than the QT in eight out of twenty-four cases. On the other hand, the AT outperforms the QT in fifteen out of twenty-four cases, and in two cases both systems perform very similarly. This is surprisingly a much better performance than expected since the AT was designed to operate with *only one* imbalance present. The presence of different imbalances for the two interfering sources meant that the AT system could not work as intended but settled to a value corresponding to some X_{imb} between that of the M-wave and the stimulus artifact. However, the duration of the two interference signals was not the same (see Fig. 5) and therefore the integrator outputs (a_1 and a_2 in Fig. 1) were mostly affected by the M-wave, which lasted longer. Consequently, the finding that the AT had similar or better performance than the QT in most of the cases, is very favourable. As the AT is currently intended for situations where only naturally occurring signals are involved and without the presence of stimulation, the AT is likely to be superior to both QT and TT. In such situations, the stimulus artefact would not be present and the interference sources would not be so close to the edges of the recording cuff as the stimulating cuff was in the described experimental setup (see Fig. 2). As a result the imbalances from different muscle signals would probably be similar. If the stimulus artefact needs to be present in other applications, it could be blanked, or a cuff with shorted outer electrodes could be used to reduce its effects [7], [11]. Even if the AT and QT perform equally well for particular imbalances, the AT has the advantage, being adaptive, of providing constant interference reduction with changing cuff conditions.

VI. Conclusion

In-vivo experiments have been performed to compare the S/I ratios of the QT, TT and AT ENG tripolar cuff recording configurations in the presence of cuff imbalance. Even though two different imbalances were present, in most cases, the AT outperformed the other two configurations based on the peak amplitude of CAP to *M*-wave. The AT may be implemented using analogue integrated circuit techniques which would result in a miniature low power implantable ENG amplifier for FES applications. It is expected that a fully integrated realization of the AT would outperform its passive counterpart described in this paper and hence, a further improvement in S/I ratio could be seen.

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