Improved spatial filtering of ENG signals using a multielectrode nerve cuff

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Abstract

In this paper we present a new multielectrode nerve cuff structure to improve the signal to noise ratio (SNR) of electroneurographic (ENG) signals. One-dimensional analysis is used to illustrate the recording of multiple signals, before a two-dimensional inhomogeneous nerve cuff model is developed. Using this model we demonstrate the effect of introducing multiple electrodes into the cuff, and the effect this has on tripolar recorded single fibre action potentials (SFAPs). Our modelled results demonstrate that an optimum number of electrodes, and therefore tripolar signals, can increase the summed SFAP peak to peak amplitude by as much as 17dB when compared to existing tripolar recording. Further theoretical analysis indicates that the action of using timing delays to align the signals cause the structure to act as a finite impulse response (FIR) frequency-sampling filter. This suggests that further improvements in ENG recording may be possible by manipulation of the cuff electrode configuration, potentially leading to selective recording by fibre size. Preliminary acute animal experiments show that the SNR improves in a √N manner, similar to an averaging process.

1. Introduction

For over 25 years, nerve cuff electrodes have proven to be a safe and reliable long-term method of electroneurographic (ENG) recording. Because ENG can be related to specific sensory information, cuff electrode recordings of ENG activity have been used in a range of applications including the upper and lower extremities [1,2] and the bladder [3,4]. Unfortunately, the low current densities produced by afferent nerve fibres, combined with the extracellular recording technique of the cuff electrode, results in an ENG signal with a very low (μV range) amplitude. In addition, interference sources either external to the body, such as mains power and electronic devices, or internal sources, often contaminate this signal. The main internal source of interference is electromyographic (EMG) activity. This *m*V amplitude signal is a result of the activity from surrounding muscle tissue, which is picked up by the electrode. Moreover, thermal and amplifier noise that is comparable to the amplitude of the ENG is also present. Therefore although theoretically promising, the practical realisation of a recording system using nerve cuff electrodes is difficult, principally due to the poor signal-to-noise ratios (SNRs) encountered. Consequently, improving the poor SNR should make the cuff signal more informative and therefore more useful.

This paper describes the evaluation of a new multielectrode nerve cuff structure, which improves the SNR of ENG recordings by using a signal averaging technique. This is similar to the approach used for characterising motor units (MUs) in surface EMG recording [5]. In order for this to work, we first consider whether it is possible to record multiple signals from within a single cuff, and the effect this has on the spatial filtering characteristics of the nerve cuff. A series of preliminary acute animal experiments then demonstrates that multiple tripolar recording is possible and that SNR improvement occurs when these signals are aligned and summed.

2. Theory

When a desired signal repeats identically at each interval, e.g. the ENG signal after each sensory stimulus, the averaging technique can satisfactorily solve the problem of separating signal from noise. Signal averaging improves the SNR by a factor √N in rms values, if the noise is uncorrelated.

In order for this method to work, each recorded signal must be identical. The action potential (AP) travelling along the nerve will be identical at each electrode, apart from a conduction velocity dependent time shift. However, the effect of the nerve cuff on the signal can not be ignored. Equation (1) shows a description of a recorded monopolar signal [6] for a narrow cuff, with the electrode not close to the cuff ends.

$$V_e(t) = c \cdot \left[\left(1 - \frac{z}{L}\right) V_m(0,t) - V_m\left(z, t - z/v\right) + \left(\frac{z}{L}\right) V_m\left(L, t - L/v\right) \right] \quad (1)$$

Where V_e is the extracellular signal, V_m is the transmembrane action potential, L is the cuff length, v is the fibre conduction velocity and $z = 0$, $z = L$ are the positions of V_m at the cuff ends; c is a constant determined by the ratio of intra- and extracellular resistances.

Equation (1) demonstrates the effect the cuff ends have on the response of the signal. In this type of recording, the extracellular signal varies depending on the position of the electrode. However, if a tripolar arrangement is used, the cuff end terms disappear, leaving only the relationship between the three electrodes as shown in equation (2).

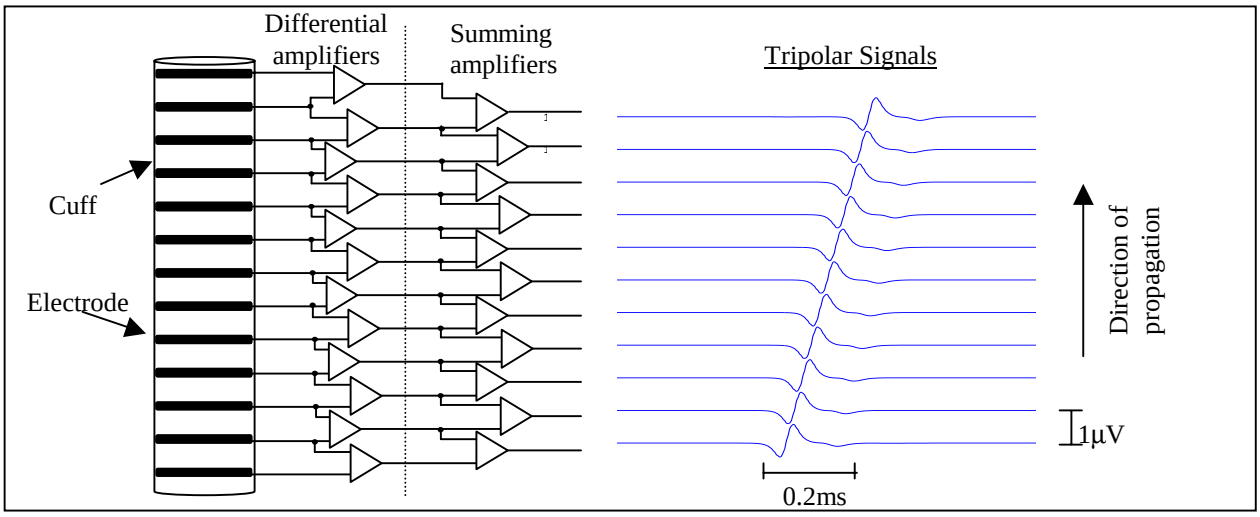

Figure 1: Tripolar signals modelled by the multielectrode cuff arrangement and parallel tripolar recording structure.

$$V_e(t) = c \cdot [V_m(z_1,t) - 2V_m(z_2, t - z_2/v) + V_m(z_3, t - z_3/v)] \quad (2)$$

Therefore, not only should it be possible to place a tripolar arrangement at any location within the cuff, but it should also be possible to record more than one tripolar signal from within the same cuff. The qualitative model used in this study consisted of two parts. The first part is a fibre model, used to calculate the membrane currents at the nodes of Ranvier along a myelinated nerve fibre, which are then regarded as current sources within a volume conductor. In the second part, a 2-dimensional inhomogeneous volume conductor model of the nerve, cuff and electrodes, as described in [7], is used to demonstrate the influence of cuff length and electrode position.

If the transmembrane action potential (V_m) is considered as an impulse (δ), then the spatial impulse response of the tripolar arrangement can be estimated by [8]:

$$h(x) = \delta(x-d) - 2\delta(x) + \delta(x+d) \quad (3)$$

Where d is the separation between impulses. Substituting the spatial positions x-d, x and x+d for electrode positions z_1, z_2 and z_3 respectively, and taking the Fourier Transform, results in equation (4).

$$H_o(\omega) = e^{-j\omega z_1/v} - 2e^{-j\omega z_2/v} + e^{-j\omega z_3/v} \quad (4)$$

Where z_1, z_2 and z_3 are the proximal, central and distal electrode positions respectively, $\omega = 2\pi f$, f is frequency and v = conduction velocity of the nerve fibre.

In the standard nerve cuff arrangement, three electrodes are evenly spaced along the length of the cuff and produce one tripolar signal. Similarly, it is fundamental to tripolar recording that this symmetry is preserved when N electrodes are employed. By connecting the electrodes in the manner shown in figure 1, the number of tripolar signals (n) recorded can be found from the relationship $n = N-2$. However, in practical terms it is better to consider the effect of cuff length and electrode separation on the number of tripoles that it is realistic to fit within a given cuff length using this parallel connection. The number of tripoles is therefore given by $n = (L/d)-1$, where L is the cuff length and d is the electrode separation.

In the frequency domain, this can be represented by multiplying equation (4) by the factors:

$$e^{-j\omega T(N-2)} \quad \text{and} \quad e^{-j\omega T(N-2)\tau}$$

representing the effect of n additional tripolar signals and the subsequent inter-tripole delay (τ) produced at the output respectively; T=d/v. In order to optimise the performance, τ is determined by correlation; or for a selective response is chosen arbitrarily. This results in equation (5), which shows the transfer function $H(\omega)$ of the multielectrode cuff, for number of electrodes N:

$$|H(\omega)| = |H_0(\omega)| \frac{\left| \sin \frac{\omega(T+\tau)(N-2)}{2} \right|}{\left| \sin \frac{\omega(T+\tau)}{2} \right|} \quad (5)$$

3. Modelling Results

Figure 1 shows a multielectrode arrangement, as well as the modelled output from each tripole. The cuff length is 30mm and contains 13 electrodes with a separation of 2.14mm (tripole length = 4.28mm). The nerve fibre illustrated is 10μm in diameter, which for 11 single fibre action potentials (SFAPs) yields an average signal of amplitude 0.96μV_{pp}. In order to establish the delay time between each tripolar output, the cross-correlation function (CCF) between the first and second output was used. Using the timing information determined by the CCF, the tripolar signals are aligned and added together. This produces a summed SFAP signal of amplitude 10.6μV_{pp}. Referring to figure 2, our results suggest that for a cuff length of 30mm, an optimum peak to peak amplitude can be achieved typically from 7 tripolar signals, produced from 9 electrodes, with a separation of 3.75mm. This produces a signal that is between 5 and 7 times greater (13.9 – 16.9dB) than a single tripolar

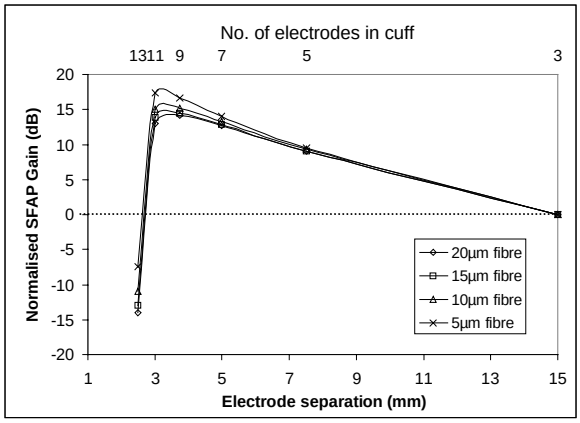

Figure 2: Normalised gain of the time shifted and summed SFAPs versus electrode separation and number, for various fibre diameters.

signal for a comparative cuff length. Correspondingly, as the rms noise component (E_{noise}) from each tripole signal will be uncorrelated, this will add in a square root manner i.e. $\sqrt{E_{noise_1} + E_{noise_2} \ldots + E_{noise_n}}$ (where n = total number of tripoles), if we assume that the noise component from each tripole will be similar, than this can be simplified as $\sqrt{nE_{noise}}$. Therefore as the signal increases by a factor n (typically up to 7 tripoles), we can estimate the overall signal to noise ratio will increase by a √n factor.

The length of the tripole determines the amplitude of each individual SFAP, which ideally should be matched to the wavelength of the largest fibre diameter to maximise the amplitude of the signal. However, the reduction of electrode separation and therefore tripole length, enables a greater number of tripolar arrangements to fit into a cuff of a particular length. Nevertheless as figure 2 shows, this affects the peak to peak amplitude of the summed SFAPs for every range of fibre diameters. Additionally, the reduced distance between each electrode increases the frequency of the SFAP, by shortening its period. This is evidently seen in the frequency response of the cuff/electrode structure presented in figure 3. This clearly shows a periodic frequency-sampling filter response that peaks at a frequency that is dependent on either the tripole length or the diameter of the nerve fibre. Plot (i-ii) of figure 3 also compare the response of a single tripole 5mm long with the multiple tripole arrangement, each of which is also 5mm long. Plots (iii-vi) of figure 3 then demonstrate the effect of varying the number of electrodes (and therefore tripoles) for two different fibre diameters. It can be seen from these plots that if the number of electrodes remains constant, the frequency response for each fibre diameter is different. This suggests that as the fibre diameter can be approximated by *5.58v* [7], the filter response can be tuned to a particular conduction velocity.

4. Experimental Methods & Results

In order to evaluate whether several tripolar signals could be recorded from with in a single nerve cuff, and

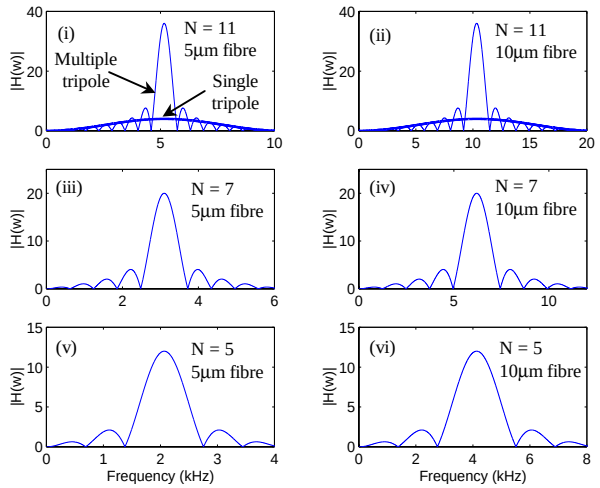

Figure 3: Frequency response plots for two different fibre diameters and N number of electrodes. The response is velocity or N dependent.

if an improvement in SNR could be achieved, a series of acute animal experiments were performed. Three New Zealand white rabbits were implanted with a new multielectrode nerve cuff. The cuff was approximately 25mm long and 1mm in diameter. It contained nine platinum electrodes, each of which is 1mm wide, symmetrically spaced 2.5mm apart. The animals were fully anaesthetised (0.3ml Hypnorm and 0.1ml Dormicum) and a spinal block (0.2ml Bupivacain) was applied between L7 and S1, to prevent reflex movements caused by any applied electrical stimulation. An incision was made on the lateral side of the left hind limb (thigh), to expose the tibial nerve above the knee. The recording cuff was placed around the tibial nerve, whereas a stimulating cuff was placed on the ramus muscularis nerve approximately 2cm proximal to the recording cuff. The lead wires exited the cuff at the distal edge and were routed percutaneously to the external amplifiers. The muscles and skin were then closed at the incision site. Tapping of the rabbit's foot sole evoked the ENG signal. The procedure was later repeated for the right leg. The data was recorded by a TEAC RD-135T, 8-channel digitizing data recorder, onto Digital Audio Tape (DAT). This recorder was always utilized in 4-channel mode, with double speed recording, enabling a sampling rate of 48k samples/sec on all channels. The recording frequency response was DC to 20kHz (+0.5 / -1 dB) under this configuration. The data was then transferred to a host computer by playing the recorded DAT tapes using the same data recorder. The resulting analogue output signals were then digitised using a 12-bit resolution, 16 channel Analogue to Digital (A/D) data acquisition card (N.I: PCI-MIO-16E-4). Unfortunately, only four channels of signal could be recorded simultaneously.

Figure 4 shows three tripolar plots of ENG recorded from the first five (proximal) electrodes. The tripoles were all recorded from the same cuff, in different locations simultaneously. The electrode separation for these recordings was 2.5mm, giving a tripole length of 5mm. The signals were bandpass filtered (300Hz-

6kHz) and amplified. The increase in SNR with the inclusion of additional tripoles is shown in figure 5, albeit for only 4 channels of signal. It can be clearly

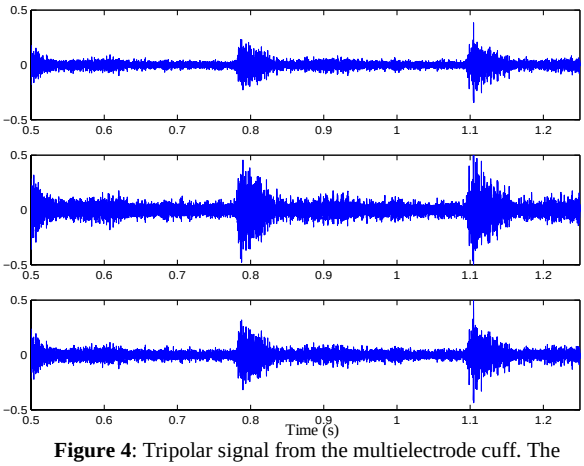

Figure 4: Tripolar signal from the multielectrode cuff. The electrode separation is 2.5mm the amplifier gain is 25000

seen that the signal to noise ratio increases in a √n manner. Also shown is the SNR response for the bipolar signals. The SNR was calculated by dividing the variance of a signal only portion of the waveform by the variance of a noise only component.

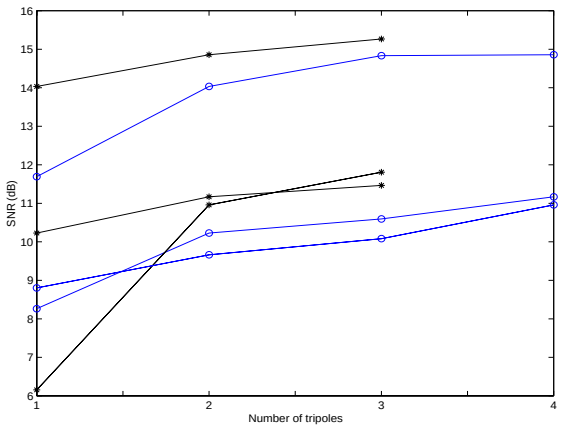

Figure 5: Improvement in signal-to-noise ratio (SNR) with increasing number of tripolar (-*-) and bipolar signals (-o-), for 3 trials

5. Discussion

The modelled results show that although the amplitude of the SFAPs is independent of tripole position within the cuff, they are dependent on electrode separation and ultimately the cuff length. Although this means that the introduction of additional electrodes effectively decreases the electrode separation, and therefore the peak to peak amplitude of each individual tripole signal, the number of tripolar signals increases. As figure 2 demonstrates, the inclusion of additional tripoles increases the SFAP amplitude in an almost linear rate up to a point where the individual tripole signals are too small to add significantly to any further increase. Figure 2 also shows that the optimal number of tripoles depends on the fibre of interest. The largest increase in signal is seen for smaller fibres. This suggests that this arrangement would be best suited to nerves containing smaller fibres, such as the bladder.

In order to establish the delay caused by the cable properties of the nerve fibre, cross-correlation can be used. Alternatively, and perhaps more practically, a variable delay time corresponding to the conduction velocity of the fibre of interest could be used. The output signal from this would relate to specific fibre velocities (diameters) and would reach a maximum value for a particular action potential velocity. The inherent delays between each tripolar output, and those introduced to allow correct summation, cause the multielectrode cuff structure to respond as a finite impulse response (FIR) filter. The inclusion of additional electrodes has the effect of not only narrowing the pass-band of the signal but also increasing its magnitude response. Also, the response is different for each fibre diameter and depends on the tripole length. This frequency-sampling filter suggests that some degree of selective recording could be achieved by variation of the electrode separation, or more practically by varying the timing coefficients. These relate directly to the conduction velocity of the nerve fibre of interest.

The primary aim of the experimental efforts was to improve the SNR of ENG signals. The effect of EMG was not considered at this time, although by using an arrangement as demonstrated in figure 1, the EMG interference can be minimised by automatic gain adjustment [8]. Future work should include a custom designed amplifier arrangement, optimised for noise performance, capable of recording a large number of channels simultaneously. This will then determine the second principle of selective fibre population recording.

6. References

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