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Fast Stimulus Artifact Recovery in a Multichannel Neural Recording System

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Abstract

We describe a 32-channel recording system and software artifact blanking technique for recording neuronal responses to high-rate electrical stimulation. Each recording channel recovers from biphasic full-scale-input pulses (1.5-V) in less than 80 μ s. Artifacts are blanked online in software, allowing flexibility in the choice of blanking period and the possibility of recovering neural data occurring simultaneously with non-saturating artifacts. The system has been used *in-vivo* to record central neuronal responses to intracochlear electrical stimulation at 2000 pulses per second. Simplicity of the hardware design makes the technique well suited to an implantable multi-channel recording system.

Index Terms

implantable recording system; neural recording; neural stimulation; stimulation artifact

I. Introduction

Many contemporary neural prostheses stimulate neural tissue using high-rate electrical pulse trains. For example, cochlear implants typically deliver biphasic current pulse trains with carrier rates at or above 1000 pulses per second (pps). These pulse trains are modulated by speech or sound signals according to a predetermined processing strategy. One method of evaluating the relative performance of various cochlear implant sound processing strategies is to record and compare midbrain neuronal responses to intracochlear stimulation in animal models [1]. However, electrical stimulation often produces artifacts that are much larger than the neuronal responses being studied. A neural recording system used in these studies would therefore benefit from a method of eliminating these artifacts. If this method were implemented in a chronic, fully implantable recording system, it would enable assessment of long-term prosthesis performance and neural plasticity resulting from chronic neural stimulation. A conceptual schematic of such a recording system is shown in Fig. 1.

Stimulus artifact removal

A number of approaches to reducing electrical stimulus artifacts have been reported previously. In the sample-and-hold method, the ability of the amplifier to follow the input signal is disabled during a stimulus pulse, and the pre-stimulus voltage is stored on a capacitor [2]. After the stimulus pulse is over, the amplifier is re-enabled. There are several drawbacks to the sample-and-hold method. First, neural activity occurring during the hold period is not recorded. This hold period cannot be changed after the recording occurs—if it is too long, excess neural information will be lost. If it is too short, the recorded waveform will be affected by stimulus artifact. Second, the hardware sample-and-hold function requires additional circuit complexity

and, therefore, increased circuit size. Third, the amplifier circuit must be passed information about when the stimulus artifact will occur, which can be impractical for an implanted amplifier that is physically separated from the stimulating device. A recent variation on the hardware sample-and-hold technique entails increasing the high-pass cutoff frequency after the hold period [3,4]. While the above drawbacks still apply to this technique, it does greatly decrease the post-artifact settling time of the system.

In an effort to overcome the disadvantages associated with the hardware sample-and-hold technique, investigators have developed algorithms for artifact reduction in software. Filtering algorithms can be used to remove the spectral components of artifacts while leaving the spectral components of neuronal action potentials intact [5,6]. These algorithms work well when the stimulus repetition rate does not fall within the frequency spectrum of the neural waveforms of interest. However, for stimulation rates between 500 and 3000 pps, the spectra of the neural waveforms and artifacts overlap, making accurate separation difficult. Furthermore, these techniques are less effective for stimulus artifact waveforms that change with time (for example, the amplitude-modulated pulse trains typically used in cochlear implants). Software-based artifact template subtraction is one approach that can be used to recover neural waveforms in the presence of spectrally-overlapping artifacts [7]. This method is also more difficult to implement for stimulus artifacts that change with time [8]. Furthermore, it is computationally expensive, making it less attractive for a multichannel recording system that removes artifacts online.

Heffer and Fallon recently described a sample-and-interpolate approach that uses a wide-bandwidth amplifier for fast artifact recovery and software interpolation for artifact removal [9]. This approach does not require additional hardware complexity, and there is no need for the recording amplifier to receive any signal indicating that a stimulus is occurring. The technique is flexible in that the artifact blanking period can be changed after recordings are made, and no neural information is lost during the stimulus artifact (provided that the amplifier does not saturate). Furthermore, this technique can effectively remove artifacts that have frequency spectra overlapping the neural signals of interest or artifacts that change with time. It is computationally efficient, making it attractive for online artifact removal.

We have developed a 32-channel recording system that uses an artifact elimination method similar to the sample-and-interpolate approach. The system is specifically designed for use in fully-implanted applications. Using this system *in-vivo* we have demonstrated the ability to record multi-channel neuronal waveforms while eliminating spectrally-overlapping stimulus artifacts.

II. Recording system design

A block diagram of the recording system headstage is shown in Fig. 2. The headstage receives inputs from a 32-channel neural probe, amplifies the neural waveforms by 100 V/V, and digitizes them with 12-bit resolution at 23.4 kSamples/s. It then sends the digitized neural signals over a serial communication link to a host PC for signal processing and display. The gain of 100 V/V combined with the ADC input range of 2.5 V gives each channel a 25-mV input dynamic range. This relatively large dynamic range prevents saturation in the case of small electrical artifacts (in our inferior colliculus recordings, artifacts from intracochlear stimulation seldom exceed 10 mV). However, even in the case of larger artifacts that do saturate the amplifier, the recording amplifiers are designed to settle rapidly. A schematic of one “fast-recovery” amplifier channel is shown in Fig. 3.

A. Signal Conditioning for fast artifact recovery

The recording amplifiers are designed to have a relatively low high-pass cutoff (1.9 Hz) and a relatively high low-pass cutoff (19.7 kHz) in order to settle quickly from electrical artifacts. Lowering the high-pass cutoff frequency reduces the magnitude of residual voltage artifacts from pulsatile electrical stimulation [10]. Large residual voltages appear as artifacts in the post-stimulus waveform, making it difficult to detect action potentials using threshold crossing methods. In some cases the residual voltage can even saturate the recording amplifier. Fig. 4 shows a numerical simulation of 100-Hz and 2-Hz RC high-pass filter responses to a biphasic stimulus pulse. The 2-Hz filter has a post-stimulus residual voltage that is over 2000 times smaller than the residual voltage of the 100-Hz filter.

Settling time from the pulse peak to the residual voltage is determined by the amplifier slew rate, low-pass cutoff frequency, and filter topology. We chose a low-pass cutoff (19.7-kHz) that is higher than typical neural recording amplifiers in order to decrease this settling time. The low-pass cutoff is set by a combination of the feedback resistor, feedback capacitor (including parasitics), and op-amp gain-bandwidth product (GBW). Approximately 1 pF of parasitic board capacitance adds to the 5 pF feedback capacitor shown in Fig. 3. The circuit was designed to allow the loosely-specified op-amp GBW of 3 MHz (Texas Instruments OPA2345) to be low enough to affect the low-pass cutoff for two reasons. First, this system is intended as a prototype for an implantable system, for which minimum power consumption is a design factor. Second, the low GBW provides a second low-pass pole for noise filtering without adding any hardware complexity. In a future design, the discrete amplifiers will be replaced by an integrated multichannel amplifier (with off-chip AC coupling capacitors) to decrease implant size and power consumption.

B. Software artifact blanking

The host PC receives a stimulus flag signal during each electrical stimulus pulse. The PC records the stimulus times and aligns them temporally with the incoming data from the recording headstage. A user-specified blanking time is added to each stimulus time to account for amplifier settling. For each recording channel, data occurring during a stimulus pulse artifact is replaced with the mean value of the data point before the stimulus and the data point after the stimulus. The resulting waveforms are copies of the originals but with flat regions wherever a stimulus artifact had been. These waveforms can then be filtered as desired in software. Since stimulus artifacts are no longer present, the high-pass cutoff can be increased and the low-pass cutoff can be decreased relative to the hardware cutoff values described above.

The hardware low-pass cutoff frequency of 19.7 kHz is above the Nyquist frequency for the specified sampling rate of 23.4 kSamples/s. For a signal passband of 3 kHz, thermal and external noise above 20.4 kHz will alias into the passband. In order to attenuate aliased external noise, we carry out recordings in an electrically-shielded recording chamber. Noise between 3 kHz and 20.4 kHz is attenuated by a 3-kHz software low-pass filter (after artifacts are removed). A higher per-channel sampling rate is possible with the analog-to-digital converters (ADCs) chosen (Analog Devices AD7490). Together, the two ADCs can sample the waveforms at up to 31 kSamples/s per channel in our design. This would be especially useful for reducing aliased thermal noise in applications that require a wider passband (e.g. an 8–10 kHz software low-pass cutoff). However, in our applications the 23.4 kSample/s sampling rate is a good tradeoff between noise performance and aggregate data rate to the host PC (12 Mb/s). Fig. 5 shows a photograph of the neural recording headstage.

III. Performance of the Recording System

Table I gives a summary of the measured performance of the recording system. When a biphasic rectangular voltage pulse (1.5-V amplitude, 40 μ s per phase, 20- μ s interphase gap) is applied directly to the amplifiers' inputs, the recording system settles within 20 μ V (RTI) of the pre-stimulus baseline voltage in a maximum of 78 μ s after the end of the pulse. For a 10-mV pulse, the settling time is a maximum of 63 μ s.

The input referred noise (100-Hz to 3 kHz) was an average of 3.6 μ V_{rms}, well below the typical noise levels of our inferior colliculus recordings (7–10 μ V_{rms}). The high-pass cutoff of the system was measured at 3.7 Hz, which is about twice the designed value of 1.9 Hz; the root of this discrepancy is a topic of ongoing investigation. However, even this measured high-pass cutoff is low enough to result in very small residual voltages (–0.5 μ V for the simulated pulse shown in Fig. 4). The average low-pass cutoff of 19 kHz is close to the designed value of 19.7 kHz. Its variation from a minimum of 17 kHz to a maximum of 21 kHz may reflect variations in op-amp GBW and parasitic board capacitance. This variation results in only a small variation in noise level (3.42 μ V_{rms} to 3.68 μ V_{rms} RTI).

The recording headstage draws 13 mA from a 3.3-V supply for a total power consumption of 43 mW. Roughly half of this power is consumed by the digital control circuitry and the other half by the amplifiers and ADCs.

In-vivo neural recordings

All procedures were approved by the UCSF Institutional Animal Care and Use Committee. A 32-channel silicon probe from NeuroNexus Technologies (Ann Arbor, MI) was inserted into the central nucleus of inferior colliculus in guinea pigs, and neural responses to intracochlear electrical stimulation were recorded (Fig. 6). The sinusoidally amplitude-modulated stimulus (1000-pps carrier) produced electrical artifacts at the input of the amplifier with a peak amplitude of about 2 mV (Fig 6a). Artifacts were removed according to the algorithm described above (Fig 6b). Stimulus pulses were programmed to be 100 μ s in duration (40 μ s/phase; 20 μ s interphase gap), suggesting that a blanking period of 175 μ s would effectively eliminate stimulus artifacts. However, the settling time of the current stimulator itself was significant, and required that the blanking period be set to 300 μ s to account for the combined settling of the current stimulator and the recording amplifier. The peaks in the post stimulus time histogram (PSTH) in Fig. 6c were delayed relative to artifact peaks by 4 – 6 ms. This delay represents synaptic and transmission latency, and would not be present if the waveforms in Fig. 6b were simply electrical artifacts. Waveforms were also recorded in response to interleaved stimulation using two intracochlear stimulus channels at 1000-pps each (not shown). Artifact rejection and neuronal responses were similar to that shown in Fig. 6, demonstrating that the system can effectively record neuronal responses to multiple cochlear implant stimulus channels at clinically-relevant carrier rates.

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Fig. 1. Schematic of implantable system for measuring midbrain responses to cochlear electrical stimulation. (1) Neural probe in target tissue. (2) Cable guide tube. (3) Implanted circuitry for amplification, filtering and analog-to-digital conversion. (4) Implanted circuitry for wireless data transmission and power reception. (5) External magnetically-aligned data receiver and power transmitter. (6) External cable connected to backpack-mounted circuitry. (Cat skull image © John Yesko, with permission.)

Fig. 2. Block diagram of headstage including fast-recovery amplifiers and analog-to-digital converters (ADCs).

Fig. 3. Circuit schematic of one fast-recovery amplifier channel, designed for a passband of 1.9 Hz to 19.7 kHz.

Fig. 4.

The effect of high-pass cutoff frequency on amplifier response to a biphasic stimulus pulse. (10-mV rectangular input pulses are 200 μ s/phase and have a 200- μ s interphase gap). (a) Decay toward baseline using a 100-Hz high-pass is considerable during the stimulus pulse phases and the interphase gap. The positive and negative decays do not sum to zero, resulting in a significant (-262μ V) residual voltage artifact following the second phase. This residual voltage artifact is large relative to most neuronal signals, and increases with stimulus pulse amplitude. (b) Decay toward baseline is considerably smaller using a 2-Hz high-pass filter, and produces a much smaller residual voltage artifact (-0.126μ V).

Fig. 5.
Photograph of the neural recording headstage.

Fig. 6.

Recordings of neuronal responses before and after stimulus artifact removal. (a) Recorded waveform includes stimulus artifacts and neuronal responses evoked by an intracochlear 50-Hz SAM pulse train (1000-pps carrier). (b) Neuronal response waveform after artifact removal (note smaller voltage scale). Horizontal line is threshold; each threshold crossing is counted as an action potential (asterisks). (c) The post-stimulus time histogram (PSTH) peaks are delayed by 4 – 6 ms relative to the artifact peaks, reflecting synaptic and conduction delays from the cochlea to the inferior colliculus. This latency confirms that the waveform in (b) is not electrical artifact.

Table I**Measured Circuit Parameters**

Parameter	Value
Settling time to 10-mV pulse ^{<i>I</i>}	63 μ s (max.)
Settling time to 1.5-V pulse ^{<i>I</i>}	78 μ s (max.)
Amplifier gain	101.2 ($\pm$ 0.3) V/V
Input referred noise (100 Hz – 3 kHz)	3.6 ($\pm$ 0.1) μ V _{rms}
High-pass cutoff (–3 dB)	3.7 ($\pm$ 0.1) Hz
Low-pass cutoff (–3 dB)	19.0 ($\pm$ 1.0) kHz

Measurements reported here include data from all 32 channels. Unless otherwise noted, values are reported as mean ($\pm$ SD).

^{*I*} Stimulus is a rectangular biphasic pulse with 40 μ s per phase and 20- μ s interphase gaps applied directly to the amplifier input. Settling time is measured from the end of the pulse until the amplifier settles within 20 μ V (RTI) of the pre-stimulus baseline voltage.