Journal of Neuroscience Methods 218 (2013) 55–71

Contents lists available at SciVerse ScienceDirect

Journal of Neuroscience Methods

journal homepage: www.elsevier.com/locate/jneumeth

Basic Neuroscience

Feedback control of electrode offset voltage during functional electrical stimulation

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HIGHLIGHTS

- A systematic design methodology was proposed to generate a feedback control system to regulate the electrode offset voltage.
- A sample-and-hold circuit was used to monitor the electrode offset voltage without interference from the stimulation current.
- A proportional–integral controller was designed based on an electrode–electrolyte interface model and a time-domain analysis.

ARTICLE INFO

Article history:
Received 8 March 2013
Received in revised form 9 May 2013
Accepted 9 May 2013

Keywords:
Electrode offset voltage
Feedback control
Functional electrical stimulation
Proportional–integral controller
Sample-and-hold circuit
Nerve cuff electrode

ABSTRACT

Control of the electrode offset voltage is an important issue related to the processes of functional electrical stimulation because excess charge accumulation over time damages both the tissue and the electrodes. This paper proposes a new feedback control scheme to regulate the electrode offset voltage to a predetermined reference value. The electrode offset voltage was continuously monitored using a sample-and-hold (S/H) circuit during stimulation and non-stimulation periods. The stimulation current was subsequently adjusted using a proportional–integral (PI) controller to minimise the error between the reference value and the electrode offset voltage. During the stimulation period, the electrode offset voltage was maintained through the S/H circuit, and the PI controller did not affect the amplitude of the stimulation current. In contrast, during the non-stimulation period, the electrode offset voltage was sampled through the S/H circuit and rapidly regulated through the PI controller. The experimental results obtained using a nerve cuff electrode showed that the electrode offset voltage was successfully controlled in terms of the performance specifications, such as the steady- and transient-state responses and the constraint of the controller output. Therefore, the proposed control scheme can potentially be used in various nerve stimulation devices and applications requiring control of the electrode offset voltage.

1. Introduction

Functional electrical stimulation has been applied in various areas of clinical research to restore damaged neural functions. The most successful examples include cochlear stimulation for hearing restoration (Wilson et al., 1991), retina stimulation for blindness (Zrenner, 2002), deep brain stimulation for Parkinson's disease (Perlmutter and Mink, 2006; Liker et al., 2008), and peripheral nerve stimulation for upper and lower limb control (Sinkjaer et al., 2003). An electrical stimulator introduces an electric charge through excitable tissue to initiate action potential through the application of either voltage- or current-controlled pulses. Voltage-controlled stimulation is more advantageous than current-controlled stimulation because it provides higher current and power efficiency with voltages closer to the supply level, leading to longer battery lifetime (Ghovanloo and Najafi, 2007). However, neither the driven current nor the injected charge is directly controlled through voltage-controlled stimulation. This drawback results in reduced degree of stimulation efficacy when the tissue properties and electrode impedance change over time, as the level of neuronal membrane depolarisation is associated with the applied current (Merrill et al., 2005). Current-controlled stimulation maintains a constant driven current throughout the pulse, and the injected charge is controlled with the duration of the pulse. Furthermore, electrode corrosion and tissue damage

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0165-0270/$ – see front matter
http://dx.doi.org/10.1016/j.jneumeth.2013.05.003

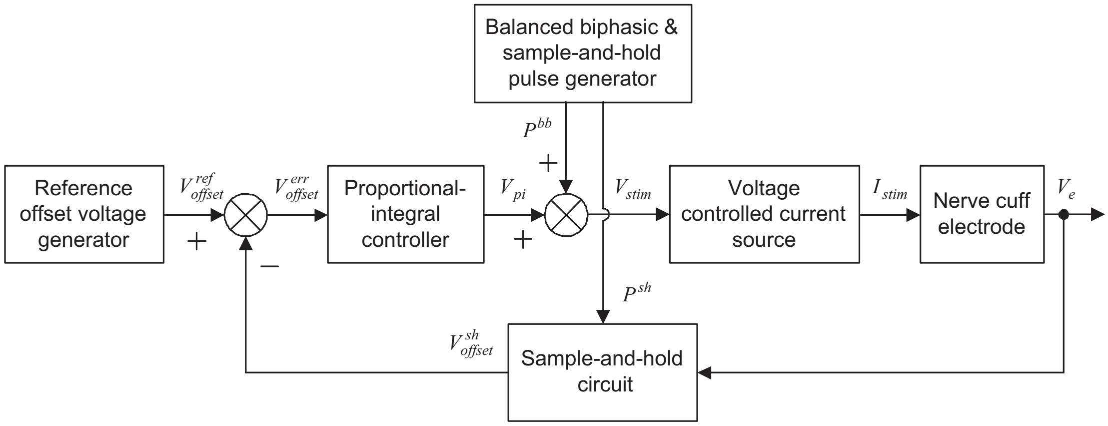

Fig. 1. Block diagram of the proposed proportional–integral controller with the sample-and-hold sensing circuit. V_{offset}^{ref} = reference offset voltage, V_{offset}^{sh} = sample-and-hold offset voltage, V_{offset}^{err} = error offset voltage, V_{pi} = output voltage of PI controller, V_{stim} = stimulation voltage, I_{stim} = stimulation current, V_e = electrode voltage, P^{bb} = balanced biphasic pulse, and P^{sh} = sample-and-hold pulse.

are prevented when a balanced biphasic current pulse is used to ensure net charge balancing at the electrode–electrolyte interface. The first pulse depolarises the neuronal membrane, thus initiating the action potential. The second pulse brings the net charge to zero, thus reversing the electrochemical processes that occur during the first pulse (Donaldson and Donaldson, 1986).

In terms of electrochemical safety, the water window is defined as the potential region across the phase boundary of the electrode within which charges are transferred between the electrode and electrolyte without causing electrolysis of the electrolyte and corrosion of the electrode (Merrill et al., 2005). In principle, the use of a charge-balanced biphasic pulse ensures that the electrode potential remains within the water window. However, in practice, a small degree of error exists in the generated pulse because of the tolerances of electronic components, and the excess charge accumulation over time drives the electrode potential beyond the water window. To resolve this problem, two approaches, namely, passive and active charge balancing, have been studied. One passive approach disposes a direct-current blocking capacitor in the path of the stimulation current (Constandinou et al., 2008; Huang et al., 1999). This method is simple and reliable to ensure zero direct current through the electrode. Generally, a blocking capacitor has a relatively large capacitance to minimise the voltage drop, and the discharge characteristics largely depend on the impedance of the electrode and the capacitor itself. Another passive approach involves periodic shorting of the electrode (Ghovanloo and Najafi, 2007; Rothermel et al., 2009). After applying a charge-balanced biphasic pulse, a shorting switch is closed to discharge any residual charge on the electrode. This method is typically combined with an additional discharge circuit to prevent large current spikes from occurring during the switching process (Sivaprakasam et al., 2005). The closing time of the switch is determined according to the time constant of the electrode and the discharge circuit. A common disadvantage of these passive approaches is that the electrode potential cannot be monitored during the discharge period; consequently, charge balancing is not guaranteed in the event of changes in the tissue properties and electrode impedance after implantation.

To overcome the disadvantage of the passive approaches, a variety of active approaches have been proposed to monitor the electrode potential during the discharge period, *i.e.*, the electrode offset voltage. One active approach involves the use of a monitoring switch to measure the electrode offset voltage after each stimulation pulse (Ortmanns et al., 2007). If the voltage exceeds a predefined water window, a short current pulse is inserted to compensate for the charge imbalance. This sequence is repeated during the non-stimulation period until the electrode offset voltage is suppressed within the water window. Nonetheless, the neuronal effects of the inserted short current pulses remain to be investigated. In a similar approach, an offset current can be applied in the background to cancel the mismatched biphasic current (Sooksood et al., 2010). The offset current is generated through the integration of the step voltage output of the water window comparator. In contrast to the pulse insertion method, the charge imbalance is not eliminated after a single instance of stimulation; however, charge balancing becomes a continuous background operation. The gain and time constant of the integrator also affect the control performance. Another active approach is to adopt a low-pass filter and a buffer for the continuous monitoring of the electrode offset voltage throughout the stimulation and non-stimulation periods (Schuettler et al., 2008). A proportional feedback controller is implemented through a non-inverting amplifier and a subtractor. The reference offset voltage can be arbitrarily changed to use the large charge injection capacity of a specific electrode material, such as iridium oxide. Despite this benefit, the steady-state error of the voltage is sensitive to the gain of the non-inverting amplifier and is difficult to remove using the proportional feedback control method. Moreover, the use of the low-pass filter and negative feedback loop causes decay in the current amplitude during the stimulation period. These active approaches have a common drawback in that the design process of the control system is heuristic and empirical for the selection of the embedding parameters, such as the amplitude and duration of the inserted pulse, the gain and time constant of the integrator, and the gain of the proportional controller.

The current study presents a systematic design methodology for a feedback control system to regulate the electrode offset voltage with an improvement in the control performance. A sample-and-hold (S/H) circuit is used to monitor the electrode offset voltage, without interference from the stimulation current. A proportional–integral (PI) controller is designed on the basis of an electrode–electrolyte interface model and a time-domain analysis. The behaviour of the PI controller is numerically simulated to guarantee charge balancing before implantation. The performance of the proposed method is evaluated through *in vitro* experiments using a nerve cuff electrode.

2. Materials and methods

A new charge-balancing system is proposed as a feedback control method of the electrode offset voltage in the current-controlled stimulation. Fig. 1 shows the block diagram of the proposed PI controller with the S/H circuit. The nerve cuff with platinum electrodes