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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.

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.

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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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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

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**Fig. 1.** Block diagram of the proposed proportional–integral controller with the sample-and-hold sensing circuit. \( V_{\text{offset}}^{\text{ref}} \) = reference offset voltage, \( V_{\text{offset}}^{\text{sh}} \) = sample-and-hold offset voltage, \( V_{\text{ref}} \) = error offset voltage, \( V_i \) = output voltage of PI controller, \( V_{\text{sim}} \) = stimulation voltage, \( I_{\text{sim}} \) = stimulation current, \( V_e \) = electrode voltage, \( p_{\text{sh}} \) = balanced biphasic pulse, and \( p_{\text{sh}} \) = sample-and-hold pulse.
is considered to be an electrode model that interfaces with the electrolyte. A voltage-controlled current source provides a bilateral current output to the electrode. The stimulation current that flows into and out of the source is changed in proportion to the stimulation voltage, which is generated as a successive pulse train with a charge-balanced biphasic pattern. The offset voltage is fed back through the S/H circuit, without interference from the stimulation current. The PI controller then adjusts the stimulation voltage to minimise the error between the reference and S/H offset voltages. In the following sections, a detailed description of each block is presented.

2.1. Nerve cuff electrode

The charge transfer at the electrode–electrolyte interface has two primary mechanisms, i.e., non-Faradaic and Faradaic reactions (Merrill et al., 2005). The non-Faradaic reactions include redistributions of the charged chemical species in the electrolyte during which no electrons are transferred between the electrode and the electrolyte. The Faradaic reactions transfer the electrons between the electrode and the electrolyte, resulting in a reduction or oxidation of the chemical species in the electrolyte. Based on these mechanisms, a simple electrical circuit model of the electrode–electrolyte interface is represented as a parallel connection of the non-Faradaic capacitance and the Faradic resistance. When working and counter electrodes are placed in the electrolyte and stimulation current passes between the electrodes through the electrolyte, electrolyte resistance appears between the two electrode–electrolyte interfaces, as shown in Fig. 2. The transfer function of the electrode is defined as the Laplace transform of the stimulation current input and the electrode voltage output through the following relationship:

\[ G_E(s) = \frac{V_E(s)}{I_{stim}(s)} = \frac{R_{Fw}}{R_{Fw}C_{nFw}s + 1} + \frac{R_c}{R_cC_{nFc}s + 1}. \]  

(1)

Here, \( R_{Fw} \) and \( R_{Fc} \) denote the Faradaic resistors, and \( C_{nFw} \) and \( C_{nFc} \) represent the non-Faradaic capacitances of the working and counter electrodes, respectively. \( R_c \) is the solution resistance.

In the current study, a cuff electrode is employed to stimulate the peripheral nerves. Fig. 3a shows that the nerve cuff was constructed on a polyimide substrate with platinum electrodes. The two end electrodes were shorted and used as the working electrode, whereas the two middle electrodes were shorted and used as the counter electrode. Each electrode was 4 mm long and 0.25 mm wide. The polyimide substrate was self-biased to curl into a roll with an inner diameter. The dimensions of the nerve cuff were optimised for application to the rat sciatic nerve. The physical characteristics of the cuff electrode are listed in Table 1. Generally, if a counter electrode has a large surface area, then it is considered to have a large capacitance and can be neglected in terms of its effect on the electrode voltage. However, in the case of the nerve cuff used here, the counter electrode has a surface area similar to that of the working electrode. Therefore, its effect should be quantified and included in the model. The impedances of the working and counter electrodes were measured using a potentiostat (VersaSTAT4, Princeton Applied Research).

![Fig. 2. Electrical circuit model of the working and counter electrodes in the electrolyte.](image)

![Fig. 3. (a) Cuff electrode for the peripheral nerve stimulation and (b) impedance magnitudes of the working and counter electrodes over the frequency.](image)

All electrochemical impedance spectroscopy measurements were performed at an open-circuit potential in a frequency range of 100 mHz to 100 kHz, with an amplitude of 10 mV. An Ag/AgCl (in saturated KCl) electrode and a platinum wire were used as reference and counter electrodes, respectively. Fig. 3b shows the impedance magnitudes of the working and counter electrodes over the frequency. The average values of the model parameters were estimated as \( R_{Fw} = 72 \) k\( \Omega \) and \( C_{nFw} = 911 \) nF for the working electrode, \( R_{Fc} = 69 \) k\( \Omega \) and \( C_{nFc} = 894 \) nF for the counter electrode, and \( R_c = 201 \) \( \Omega \). The small difference between the estimated model parameters for the working and counter electrodes might reflect experimental artefacts and manufacturing variations. From this measurement, we observed that the Faradic resistor and the non-Faradic capacitor of the working electrode had values similar to those of the counter electrode. Accordingly, the transfer function of the nerve cuff electrode is simplified using the following equation:

\[ G_E(s) = \frac{V_E(s)}{I_{stim}(s)} \approx \frac{R_{Fw}R_c}{R_{Fw}C_{nFw}s + R_c + 2R_{Fw}}. \]

(2)

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Measured value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Diameter of cuff</td>
<td>1 mm</td>
</tr>
<tr>
<td>Thickness of cuff</td>
<td>18 ( \mu )m</td>
</tr>
<tr>
<td>Distance between end electrodes</td>
<td>3 mm</td>
</tr>
<tr>
<td>Distance between middle electrodes</td>
<td>0.5 mm</td>
</tr>
<tr>
<td>Length of electrode</td>
<td>4 mm</td>
</tr>
<tr>
<td>Width of electrode</td>
<td>0.25 mm</td>
</tr>
</tbody>
</table>

Table 1

Physical characteristics of the cuff electrode.
2.2. Voltage-controlled current source

When either end of the electrode is grounded, as shown in Fig. 2, a voltage-controlled current source for a grounded load is commonly used to provide the stimulation current through the electrode into the tissue. In this case, the output impedance must be sufficiently large, and the output current error should be minimised over a wide range of stimulation frequencies. To this end, a dual-operational amplifier current source (Dickin and Wang, 1996), as shown in Fig. 4, is adopted. This circuit is based on the positive feedback architecture, and the voltage drop across $R_{\text{stim}}$ is maintained as a constant value. The relationship between the stimulation voltage and the stimulation current is expressed using the following equation:

$$C_{e}(s) = \frac{I_{\text{stim}}(s)}{V_{\text{stim}}(s)} = \frac{1}{R_{\text{stim}}}.$$  \hspace{1cm} (3)

The supply voltages of the amplifiers, namely, VDD and VSS, were 15 V and -15 V, respectively. These components are not drawn in the schematic for clarity. The resistor $R_{\text{stim}}$ was set to 1 kΩ, i.e., the 0.5 V stimulation voltage was converted to a 500 μA stimulation current.

2.3. Sample-and-hold circuit

The main purpose of this study is to introduce the concept of feedback control to the charge-balancing problem. To construct the feedback loop, the electrode offset voltage must be monitored continuously, not intermittently. When the stimulation current is applied to the electrode, a large peak-to-peak voltage is observed, and its offset voltage cannot be easily measured using conventional filter techniques. As an alternative solution, the S/H scheme is proposed to sense the electrode offset voltage. When the stimulation waveform is given as a biphasic pulse with a repetition frequency, a synchronised monophasic pulse with a stimulation period is applied to the S/H circuit. The electrode voltage is subsequently captured at the beginning of the monophasic pulse and maintained until the end of the monophasic pulse. Otherwise, the electrode voltage passes through the S/H circuit, without distortion. Because feedback control is maintained throughout the stimulation and non-stimulation periods, it is possible to neglect the change in the electrode offset voltage that occurs during the stimulation period. As a result, this scheme is free from the interference of the successive stimulations, facilitating continuous sensing of the electrode offset voltage.

The S/H circuit is implemented using analogue switches and buffers, as shown in Fig. 5a. During the sample mode, the switches are closed, and the output voltage follows the input voltage. In the hold mode, the switches are opened, and the input voltage is maintained through holding capacitors $C_h$. The pedestal error and hold-time glitch are reduced through a compensation network $R_c$ and $C_c$.

![Fig. 5. (a) Schematic of the sample-and-hold circuit and (b) typical sample-and-hold operation.](image-url)
Fig. 6. (a) Schematic of the balanced biphasic and sample-and-hold pulse generator and (b) typical pulse generation.

59. Fig. 5b illustrates a typical S/H operation. When an imbalanced biphasic current pulse is applied to the electrode, the electrode voltage does not vanish at the end of the single pulse, and its offset voltage gradually increases according to the successive pulses. The electrode voltage is then sampled and maintained at the timing of the S/H pulses. As a result, the S/H offset voltage is given as

\[ V_{\text{offset}}(t) = \begin{cases} V_e(t_1 + kT) & t_1 + kT \leq t < t_2 + kT \\ V_e(t) & \text{otherwise} \end{cases} \]

where \( k = 0, 1, 2, \ldots \) and \( t_1, t_2, \) and \( T \) denote the beginning and ending times of the S/H pulse and the repetition period, respectively.

To design the PI controller, we assume that the reference offset voltage is given by a step function and the balanced biphasic pulse is not applied to the electrode because the purpose of the controller is to regulate the electrode offset voltage, rather than the electrode voltage; the balanced biphasic pulse is considered to be a disturbance. In this case, the electrode voltage is approximated through the S/H offset voltage. Consequently, the transfer function of the S/H circuit is expressed using the following equation:

\[ G_{\text{sh}}(s) = \frac{V_{\text{offset}}(s)}{V_e(s)} \approx \frac{V_{\text{offset}}(s)}{V_{\text{offset}}(s)} = 1 \text{ when } P^{\text{sh}}(s) = 0. \]
2.4. Balanced biphasic and sample-and-hold pulse generators

As noted in the previous section, the biphasic pulse consists of the cathodic and anodic pulses, and the S/H pulse has a monophasic waveform, with a width identical to that of the biphasic pulse. These pulses are generated using timers and differential amplifiers, as shown in Fig. 6a. Two timers are configured in series to perform monostable operations. Fig. 6b illustrates a typical pulse generation. The negative trigger pulse is applied through an external device with a width less than that of the cathodic pulse. The cathodic pulse is fired from the falling edge of the trigger pulse and is connected to a high-pass filter. The anodic pulse is subsequently fired from the falling edge of the high-pass filter output. The widths of the cathodic and anodic pulses are controlled through the combination of the resistor $R_t$ and capacitor $C_t$, calculated using the following equation:

$$\frac{t_2 - t_1}{2} = 1.1 \cdot R_t C_t. \quad (6)$$

The outputs from the timers are subtracted using a difference amplifier, and the inverted cathodic pulse is therefore added to the anodic pulse. The resulting biphasic pulse is fed through a voltage divider and a voltage follower to adjust the amplitude. In addition, the S/H pulse is produced with the same cathodic and anodic pulses used for the biphasic pulse. The outputs from the timers are added using an inverting summing amplifier, and their polarity is reversed using an inverting amplifier. This circuit synchronises the S/H pulse with the biphasic pulse. The supply voltage VCC of the timers was 5 V. The resistor $R_t$ and capacitor $C_t$ were set to 9.1 kΩ and 0.025 µF, respectively. The potentiometer POTs, was set to a position that connected the centre tap by 1111 Ω to the ground. The repetition frequency and width of the negative trigger pulse were set to 50 Hz and 50 µs, respectively. As a result, the amplitudes and widths of the cathodic and anodic pulses were determined as 0.5 V and 250 µs, respectively. The S/H pulse was generated with an amplitude of 5 V and a width of 500 µs.

2.5. Reference offset voltage generator

The reference offset voltage generator is implemented using a voltage divider and a voltage follower, as shown in Fig. 7. The reference offset voltage is defined by the setting of the potentiometer. A 10 kΩ resistance between the centre tap of the potentiometer and the VSS generates a zero reference offset voltage. Cogan et al. (2006) used potential biasing and biphasic current pulse waveforms to maximise the charge injection capacity of the activated iridium oxide microelectrodes. In our proposed charge-balancing system, potential biasing is easily achieved through changing the centre-tap position of the potentiometer. For example, a reference offset voltage of 400 mV is obtained when the centre tap is connected through 10,548 Ω to the VSS.

2.6. Proportional–integral controller

The remaining part of the proposed charge-balancing system is the PI controller. The purpose of the controller is to drive the S/H offset voltage to track the given reference offset voltage. Thus, the error offset voltage is defined using the following equation:

$$V_{err}^o(t) = V_{ref}^o(t) - V_{off}^o(t). \quad (7)$$

The PI controller produces an output proportional to the current error and the accumulated error over time (Ogata, 1997):

$$V_p(t) = K_p \int_0^t V_{err}^o(t) \, dt + K_i \int_0^t V_{err}^o(t) \, dt. \quad (8)$$

Here, $K_p$ and $K_i$ are the proportional and integral gains, respectively. The stimulation voltage is then constructed through the summation of the output of the PI controller and the balanced biphasic pulse:

$$V_{stim}(t) = V_p(t) + \rho_{BB}(t). \quad (9)$$

As noted in Section 2.3, the balanced biphasic pulse is considered to be a disturbance and is not applied to the electrode during controller design. Therefore, the stimulation voltage is equal to the output of the PI controller, and the transfer function of the PI controller is expressed using the following equation:

$$G_{pi}(s) = \frac{V_p(s)}{V_{off}^o(s)} = \frac{V_{stim}(s)}{V_{off}^o(s)} = K_p + \frac{K_i}{s} \quad \text{when } \rho_{BB}(s) = 0. \quad (10)$$

Using Eqs. (2), (3), (5), and (10), the transfer-function block diagram of the proposed charge-balancing system is presented, as shown in Fig. 8. Because the system has a unity feedback of $G_{db}(s) = 1$, the closed-loop transfer function is established via

$$\frac{V_{off}^o(s)}{V_{off}^o(s)} = \frac{G_{pl}(s)G_{cs}(s)G_{tc}(s)}{1 + G_{pl}(s)G_{cs}(s)G_{tc}(s)} \quad (11)$$
where the open-loop transfer function is expressed using the following equation:

\[ G_{pl}(s)G_{e}(s)G_{c}(s) = \frac{(K_{p}s + K_{i})(R_{p}R_{f}C_{f}wF_{w}s + R_{s} + 2R_{f}w)}{R_{stim}^{s}(R_{f}C_{f}wF_{w}s + 1)}. \]  

(12)

For charge-balancing problems, the reference offset voltage is given by a step function with an arbitrary magnitude. In the current study, assuming a unit-step function, the specifications of the control performance are set to the following values:

1. Steady-state error = 0,
2. Maximum overshoot ≤ 12%,
3. Rise time ≤ 1.5 ms,
4. Settling time ≤ 6 ms,
5. Maximum output of the controller ≤ 1000 μA.

The steady-state response due to the unit-step input is evaluated as specified (1). The transient-state response due to the unit-step input is assessed as specifications (2)-(4), where the maximum overshoot is defined as 100 times the maximum peak value of the step response minus its final value divided by its final value. The rise time is defined as the time required for the step response to rise from 10% to 90% of its final value. The settling time is defined as the time required for the step response to decrease and remain within 5% of its final value. In addition, the controller output is constrained by specification (5), which minimises the neuronal effects of the controller output during the compensation of the charge imbalance.

In terms of the pole-zero configuration, the immediate effect of the PI controller is the addition of a pole at \( s = 0 \) to the open-loop transfer function. When the reference offset voltage is a unit-step function, the steady-state error is defined using the following equation:

\[ e_{ss} = \lim_{t \to \infty} V_{offset}^{err}(t) = \lim_{s \to 0} V_{offset}^{err}(s) = \lim_{s \to 0} sV_{offset}^{err}(s) = \left(1 + G_{pl}(s)G_{e}(s)\right)^{-1}. \]  

(13)

Thus, the PI controller reduces the steady-state error due to the unit-step input to zero.

Accordingly, the problem with the design of the PI controller is simplified to choose the proportional and integral gains that satisfy the given specifications of the transient-state response and the controller output. Because the PI controller is essentially a low-pass filter, the closed-loop system usually has a slower rise time and longer settling time. However, if the zero position at \( s = -K_{i}/K_{p} \) is properly selected, the transient-state response is improved. The current study uses root-locus techniques to propose a systematic method for designing the PI controller. In the linear control theory, the root-locus techniques facilitate the investigation of the trajectories of the roots of the characteristic equation when a certain system parameter varies (Kuo, 1995). The characteristic equation of the closed-loop system is obtained by setting the denominator polynomial of the transfer function to zero. Thus, the roots of the characteristic equation must satisfy the following equation:

\[ 1 + G_{pl}(s)G_{e}(s)G_{c}(s) = 0. \]  

(14)

The roots of the characteristic equation are the poles of the closed-loop transfer function and determine the stability and transient-state response of the closed-loop system. To evaluate the control performance with the PI gains, we assume that the open-loop transfer function contains a variable parameter \( K \) as a multiplying factor. Therefore, the characteristic equation is rewritten as the following equation:

\[ 1 + K G_{pl}(s)G_{e}(s)G_{c}(s) = 1 + K \left(\frac{(K_{p}s + K_{i})(R_{p}R_{f}C_{f}wF_{w}s + R_{s} + 2R_{f}w)}{R_{stim}^{s}(R_{f}C_{f}wF_{w}s + 1)}\right) = 0. \]  

(15)

In this equation, it is clear that when \( K = 0 \), the poles of the open-loop transfer function are the roots of the characteristic equation; when \( K = \infty \), the zeros of the open-loop transfer function are the roots of the characteristic equation. These root-locus properties are useful for efficiently selecting the proper PI gains and for understanding the control performance of the closed-loop system by varying parameter \( K \) from 0 to \( \infty \).

The roots of the characteristic equation should be located in the left-half \( s \)-plane to satisfy the stability criterion of the linear control system. From the relationship between the characteristic equation and the open-loop transfer function, the zero position at \( s = -K_{i}/K_{p} \) should have a value less than zero. To select the position of the zero, the pole-zero configuration of the open-loop transfer function is investigated, as shown in Fig. 9. The open-loop transfer function that satisfies Eq. (15) can be rewritten as the following equation:

\[ G_{pl}(s)G_{e}(s)G_{c}(s) = \left(\frac{(K_{p}s + K_{i})(R_{p}R_{f}C_{f}wF_{w}s + R_{s} + 2R_{f}w)}{R_{stim}^{s}(R_{f}C_{f}wF_{w}s + 1)}\right) = -\frac{1}{R_{f}C_{f}wF_{w}}. \]  

(16)

As parameter \( K \) approaches zero, \( G_{pl}(s)G_{e}(s)G_{c}(s) \) approaches infinity; thus, the roots of the characteristic equation must approach the pole of the PI controller at \( s = 0 \) and the pole of the electrode model at \( s = -(R_{f}C_{f}wF_{w})/(R_{p}R_{f}C_{f}wF_{w}) \). Similarly, as parameter \( K \) approaches infinity, \( G_{pl}(s)G_{e}(s)G_{c}(s) \) approaches zero; thus, the roots of the characteristic equation must approach the zero of the PI controller at \( s = -K_{i}/K_{p} \) and the zero of the electrode model at \( s = -(R_{f}R_{f}C_{f}wF_{w})/(R_{p}R_{f}C_{f}wF_{w}) \). From this pole-zero configuration, the locations of the pole and the zero of the electrode model separate the possible regions for the position of the zero of the PI controller. The numerical simulation results showed that a zero located close to or within Region 1 provides an over-damped response in a transient state; in addition, a zero located close to or within Region 3 generated a large controller output that exceeded the power supply voltages. According to these results, the position of the zero was in Region 2, thus satisfying the following condition:

\[ \frac{R_{f} + 2R_{f}w}{R_{f}R_{f}C_{f}wF_{w}} < -\frac{K_{i}}{K_{p}} < -\frac{1}{R_{f}C_{f}wF_{w}}. \]  

(17)

As a trade-off choice, the position of the zero was set at \( s = -400 \). Fig. 10 shows the root loci of Eq. (15) with \( -K_{i}/K_{p} = -400 \), when \( K \) varied continuously from 0 to \( \infty \). It was shown that the complex portion of the root loci was a circle and that the two breakaway points were located on the real axis. The addition of the zero to the open-loop transfer function moved and bent the root loci towards
Fig. 10. Root loci of the characteristic equation with \( -K_p/K_r = -400 \).

The schematic of the PI controller is shown in Fig. 12, where the proportional and integral terms are implemented using an inverting amplifier and an integrator, respectively. The negative gains of the proportional and integral terms are reversed to positive gains using an inverting summing amplifier. The error offset voltage is generated using a difference amplifier, and its negative sign is cancelled using an additional difference amplifier, which is used to add the output of the PI controller to the balanced biphasic pulse. As a result, the output of the PI controller is given as the following equation:

\[
V_{\text{out}}(t) = V_p(t) + V_i(t) = \frac{R_p2}{R_p1} V_{\text{err}}(t) + \frac{1}{R_iC_i} \int_0^t V_{\text{err}}(\tau) d\tau. \tag{19}
\]

A comparison of Eqs. (8) and (19) revealed that the parameters of the PI controller are associated with the circuit parameters, as define in the following equation:

\[
K_p = \frac{R_p2}{R_p1}, \quad K_i = \frac{1}{R_iC_i}. \tag{20}
\]

For the value of the proportional gain in Eq. (18), the resistors \( R_{p1} \) and \( R_{p2} \) were chosen as 10 k\( \Omega \) and 7.57 k\( \Omega \), respectively. As previously described, the integral gain is inversely proportional to the value of the capacitor of the integrator. Thus, the large integral gain in Eq. (18) corresponded to a small capacitor, and the resistor \( R_i \) and capacitor \( C_i \) were set to 3.29 k\( \Omega \) and 1000 nF, respectively.

2.7. Performance evaluation

The charge-balancing performance of the proposed method was evaluated for different pulse waveforms with zero or non-zero

The left-half s-plane. Therefore, if the \( K \) value is selected properly, the control performance of the closed-loop system is improved. Fig. 11 shows the unit-step responses and outputs of the proposed PI controller corresponding to several \( K \) values plotted in Fig. 10. From these results, we observed that, for all cases, the steady-state error was reduced to zero, satisfying specification (1). In addition, when the \( K \) value increased, an improvement in the transient-state response was observed, whereas an increment in the controller output was required. The influence of parameter \( K \) on the control performance is quantified in Table 2, where the measured values are shown along with the roots of the characteristic equation for various \( K \) values. The results in Table 2 confirmed that when \( K \geq 0.1380 \), the maximum overshoot, rise time, and settling time is further reduced compared with those of specifications (2)-(4). However, when \( K = 0.2966 \), the controller output was more excessive than that of specification (5). Accordingly, the optimal value of \( K \), which satisfies the specifications, was approximately 0.15.

Selecting \( K = 0.1515 \) with \( K_p = 2000 \) and \( K_r = 5 \), the following results were obtained for the parameters of the PI controller using Eq. (15):

\[
\begin{align*}
K \cdot K_p & \rightarrow K_p = 0.7575, \\
K \cdot K_r & \rightarrow K_r = 303.
\end{align*} \tag{18}
\]

The integral gain \( K_i \) was selected to have a larger value than that of the proportional gain \( K_p \) because this value is inversely proportional to the value of the capacitor in the implementing circuit. For practical reasons, it is necessary to watch for an unrealistically large capacitor value.

![Figure 10](image-url)
reference offset voltages. The PI controller was set with gains $K_p = 0.7575$ and $K_i = 303$. The stimulation current and electrode voltage were measured using two instrument amplifiers (INA103, Texas Instruments), which offer a low noise and a wide bandwidth. The voltage drop across resistor $R_{	ext{trim}} = 1 \, \text{k} \Omega$ was amplified, and the corresponding stimulation current was calculated, whereas the electrode voltage was detected at the working electrode with reference to the ground. All experiments were performed in vitro using a nerve cuff electrode immersed in a saline solution (0.9% NaCl) at room temperature.

2.7.1. Charge-balancing performance for biphasic pulses

The proposed method was verified for the balanced biphasic pulses with a zero reference offset voltage. The balanced biphasic and S/H pulses were produced through the proposed pulse generator using the method described in Section 2.4. The amplitudes and widths of the cathodic and anodic pulses were determined as 0.5 V and 250 µs, respectively. The S/H pulse was generated with an amplitude of 5 V and a width of 500 µs. A function generator (AFG3022B, Tektronics) was used to provide the negative trigger pulse train with a repetition frequency of 50 Hz and a width of 50 µs.

The following experiment was performed to investigate the ability of the potential biasing across the electrodes. The reference offset voltage was generated as described in Section 2.5 and set to 400 mV. The balanced biphasic and S/H pulses were identical to those used in the previous experiment.

2.7.2. Charge-balancing performance for monophasic pulses

The purpose of this experiment was to examine the response of the proposed method to the monophasic pulses with a zero reference offset voltage. Monophasic pulses are considered to be an extreme case of charge imbalance. The monophasic and S/H pulses could also be generated through the removal of the high-pass filter from the proposed pulse generator. The amplitude and width of the cathodic pulse were determined as 0.5 V and 250 µs, respectively.

![Fig. 11. (a) Unit-step responses and (b) outputs of the PI controller for various $K$ values.](image-url)
2.7.3. Effects of the sample-and-hold circuit
To evaluate the effects of the S/H circuit on the charge-balancing performance, the responses to the balanced biphasic pulses were examined with only the sample operation or the S/H operation. When only the sample operation was applied, the switch input to the S/H circuit was disconnected from the pulse generator and grounded, thereby facilitating the continuous monitoring of the electrode voltage, without distortion throughout the stimulation and non-stimulation periods. The balanced biphasic and S/H pulses were identical to those used in Section 2.7.1, and the reference offset voltage was set to zero.

2.7.4. Effects of the proportional–integral controller
To assess the effects of the PI controller on the charge-balancing performance, the responses to the monophasic pulses were investigated using several values of the PI gains. The monophasic and S/H pulses were identical to those used in Section 2.7.2, and the reference offset voltage was set to 400 mV.

The charge-balancing performance of the proposed PI controller was compared with that of the P controller. For this experiment, the P controller was implemented through the elimination of the integrator from the proposed PI controller. The monophasic and S/H pulses were identical to those used in Section 2.7.2, and the reference offset voltage was set to 400 mV.

2.7.5. Robustness of the proportional–integral controller against tissue growth
To verify the influence of the tissue growth on the charge-balancing performance, a saline-tank experiment was performed, which was modified from previous studies (Triantis and Demosthenous, 2008; Chu et al., 2012). Fig. 13 shows a nerve cuff immersed in a saline solution (0.9% NaCl) and connected to the proposed stimulator. To present a tissue resistance through the tissue growth inside the cuff, a cylinder-shaped obstacle (1-mm diameter and 10-mm long) was created using silicon on a Teflon-coated stainless steel wire and placed inside the cuff. The responses of the proposed method to the monophasic pulses were investigated when the obstacle was outside or inside the cuff. The monophasic and S/H pulses were identical to those used in Section 2.7.2, and the reference offset voltage was set to zero.

3. Results and discussion
3.1. Charge-balancing performance for biphasic pulses
Fig. 14 shows the response to the balanced biphasic pulses with a zero reference offset voltage. During the stimulation period, the 0.5 V stimulation voltage was converted to a 500 μA stimulation current according to the relationship shown in Eq. (3). Meanwhile, throughout the stimulation and non-stimulation periods, the electrode offset voltage was monitored using the S/H circuit, without interference from the stimulation current. Although the biphasic pulse was designed to be charge balanced, a small charge imbalance existed in the generated pulse because of the tolerances of the electronic components. As a result, the electrode voltage did not vanish at the end of each single pulse, and the corresponding offset voltage with a peak value of approximately 3.2 mV was observed in the output of the S/H circuit. To compensate for this
Fig. 14. Response to balanced biphasic pulses with a zero reference offset voltage.

Fig. 15. Response to balanced biphasic pulses with a non-zero reference offset voltage.
charge imbalance, the PI controller generated a counter-phase voltage, which was converted to a negative stimulation current, and the offset voltage was regulated as 0 V within the non-stimulation period. The measured values for the control performance were as follows: steady-state error of 0 V, maximum overshoot of 6.7%, rise time of 0.73 ms, settling time of 2.4 ms, and maximum controller output of 2.6 µA. Consequently, these results satisfied the control performance specifications shown in Section 2.6.

Fig. 16. Response to monophasic pulses with a zero reference offset voltage.

Fig. 17. Response to monophasic pulses with a zero reference offset voltage when the feedback control was not applied.

Fig. 15 shows the response to the balanced biphasic pulses with a non-zero reference offset voltage. During the initial period before the application of the pulse, the error between the reference and S/H offset voltages was quickly reduced through the PI controller. This initial response was evaluated in terms of the control performance: steady-state error of 0 V, maximum overshoot of 11.5%, rise time of 1.1 ms, settling time of 5.6 ms, and maximum controller output of 245 µA. In addition, the charge imbalance due to the biphasic
pulse was offset in the same manner used in the previous experiment. Consequently, the offset voltage was maintained at 400 mV over time, satisfying the predetermined control performance specifications.

3.2. Charge-balancing performance for monophasic pulses

Fig. 16 shows the response to the monophasic pulses, with a zero reference offset voltage. After each monophasic pulse, a peak offset voltage of approximately −0.19 V was observed in the output of the S/H circuit. To prevent excess charge accumulation, the PI controller generated a counter-phase voltage, which was converted to a positive stimulation current and acted as an anodic pulse. The steady- and transient-state responses and the controller output met the control performance specifications: steady-state error of 0 V, maximum overshoot of 11.8%, rise time of 1.0 ms, settling time of 5.9 ms, and maximum controller output of 142 µA.

Fig. 17 shows the response to the monophasic pulses, with a zero reference offset voltage, when the feedback control was not applied. Immediately after the first monophasic pulse occurred, the electrode voltage increased to −0.31 V. The electrode was subsequently discharged through the Faradaic resistances of the electrode–electrolyte interface, and the electrode voltage leaked off to −0.073 V at 40 ms. However, the second monophasic pulse recharged the electrode with a shorter time constant than that of the discharge. Therefore, the electrode voltage again increased to −0.39 V. Consequently, successive monophasic pulses accumulated the residual charge on the electrode over time and shifted the electrode voltage to a more negative value.

3.3. Effects of the sample-and-hold circuit

Fig. 18a shows the response to the balanced biphasic pulses with a zero reference offset voltage when only the sample operation was applied. The output of the S/H circuit was identical to the electrode voltage. Thus, during the stimulation period, the error offset voltage occurred with considerable peak-to-peak variation, and the corresponding controller output affected the amplitude of the stimulation current. As shown in the plot of the stimulation current, the cathodic current decayed to 59.5% of the initial value after 250 µs. Subsequently, the anodic current increased initially to 686 µA, followed by a decrease to 59.5% of its initial value after 250 µs. This effect resulted in an increase in the electrode voltage during the non-stimulation period compared with the case when the S/H operation was applied.

It is important that the amplitude of the stimulation current is maintained throughout the pulse because the neuronal membrane depolarisation and the reverse electrochemical process are related to the applied current. To satisfy this requirement, the proposed method was designed to generate a constant current during the
stimulation period and control the electrode voltage to a pre-determined value during the non-stimulation period. As shown in Fig. 18b, during the stimulation period, the electrode voltage was maintained through the S/H circuit, and the controller output did not affect the amplitude of the stimulation current. During the non-stimulation period, the electrode voltage was sampled by the S/H circuit and was rapidly controlled to the zero reference offset voltage.

3.4. Effects of the proportional–integral controller

Fig. 19 shows the responses of the PI controller, corresponding to several $K$ values, as shown in Fig. 11. In all cases, a peak electrode voltage of approximately 0.07 V was observed after a monophasic pulse. As the gain increased, the maximum overshoot decreased, and the rise and settling times decreased. Meanwhile, the maximum output of the controller increased. The controller with the
lowest gain had a larger settling time than the non-stimulation period. Hence, the electrode offset voltage did not converge to the predetermined reference value before the next stimulation pulse. When the electrode offset voltage was controlled with a high gain, the controller generated a large stimulation current at the beginning of the non-stimulation period, which quickly eliminated the error offset voltage. A further increase in the gain resulted in a more rapid settling of the electrode offset voltage down to 400 mV, and the stimulation current more highly peaked and then sharply diminished. These experimental results were consistent with the simulation results shown in Fig. 11, indicating that the control performance of the electrode offset voltage could be improved through proper selection of the PI gain.

Fig. 20 shows the response of the P controller when $K = 0.1515$. The transient-state response and the controller output met the control performance specifications. However, the steady-state error did not vanish but remained constant at $-0.03$ V. This result indicated that the integral term of the proposed PI controller reduces the steady-state error due to the step input to zero.

### 3.5. Robustness of the proportional–integral controller against tissue growth

Fig. 21a shows the response to the monophasic pulses with a zero reference offset voltage when the obstacle was outside the cuff to simulate a case without tissue growth. These results were identical to those shown in Fig. 16. When the obstacle was positioned to occupy the interior of the cuff to simulate a case with tissue growth, the electrode voltage and stimulation current are shown in Fig. 21b. After a monophasic pulse, a peak electrode voltage of approximately $-0.55$ V was observed, and the peak electrode voltage increased to almost two times this value when the obstacle was outside the cuff. Nonetheless, the steady- and transient-state responses and the controller output met the control performance specifications: steady-state error of 0 V, maximum overshoot of 10.5%, rise time of 0.8 ms, settling time of 6.0 ms, and maximum controller output of 114 $\mu$A. These comparative results indicated that the proposed PI controller was robust against tissue growth inside the cuff.

In vivo, the estimated equivalent circuit parameters of the electrode model might change over time due to tissue growth around the electrodes and protein absorption on the electrodes. These factors can be modelled as a tissue resistance $R_t$ between the two electrode–electrolyte interfaces, as shown in Fig. 22a. To show the influence of the tissue resistance, a portion of Fig. 21b is enlarged in Fig. 22b. $V_1$ is the voltage across the electrolyte and tissue resistances at the beginning and end of the monophasic pulse, with an amplitude of $I_1$, i.e., $V_1 = I_1 (R_t + R_s)$, whereas $V_2$ is the voltage across the two electrode–electrolyte interfaces during the stimulation period. From this measurement, the value of the tissue resistance was estimated as $R_t = 402 \, \Omega$, and the pole-zero configuration of the open-loop transfer function and the root loci of the characteristic equation were modified, as shown in Fig. 23. The addition of the tissue resistance influenced the movement of the zero of the electrode model at $s = -(R_t + R_s + 2R_{fw})/(R_t + R_s)RFwCnFw$ towards the right. The circle representing the complex portion of the root loci became smaller than that without the tissue resistance, and
the breakaway point between the zeros of the electrode model and the PI controller was shifted to the right. In addition, when $K = 0.1515$, the roots of the characteristic equation were moved to the right along the circle. However, the change in the parameters of the electrode model did not induce a large variation in the roots of the characteristic equation. The roots corresponding to $K = 0.1515$ remained in the neighborhood of the optimal root locations specified in Table 2. Consequently, the control performance of the closed-loop system was insensitive to the change in the parameters of the electrode model due to tissue growth.

4. Conclusions

This paper proposed a new feedback control scheme to regulate the electrode offset voltage in functional electrical stimulation. The proposed scheme comprises an electrode, a voltage-controlled current source, a balanced biphasic pulse generator, an S/H circuit, an S/H pulse generator, a reference offset voltage generator, and a PI controller. To construct a feedback loop, the S/H circuit and S/H pulse generator provided continuous monitoring of the electrode offset voltage, without interference from the stimulation current. The change in the electrode offset voltage during the stimulation period was neglected, whereas the electrode offset voltage during the non-stimulation period was fed back to the PI controller. This configuration simplified the design problem of the PI controller to one involving a unity feedback loop. Based on the electrode–electrolyte interface model, the proportional and integral gains were selected to satisfy the given control performance specifications. From the experimental results, the amplitude of the stimulation current during the stimulation period was maintained at a constant level equal to the output of the balanced biphasic pulse generator. In addition, the performance of the PI controller for different pulse waveforms confirmed that the proposed feedback control scheme was effective in solving the charge-balancing problem.

Although the charge-balancing performance of the proposed control scheme was evaluated in vitro using a nerve cuff electrode, further safety studies should be performed in vivo to examine clinical functional electrical stimulation. In future work, the proposed circuitry will be miniaturised for extension to a multi-channel system for simultaneous stimulation at multiple sites.

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References


Fig. 23. Root loci of the characteristic equation with the tissue resistance.


