A Precise Active Charge Balancing Method for Neural Stimulators by Utilizing Polarity Changes of the Remaining Voltage

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Abstract

A new precise active charge balancing method for safe electrical stimulation has been presented in this paper. This method can be adopted in a stimulator to be used for neural electrical stimulation. After each stimulation the electrode voltage is monitored by a dynamic comparator. The output of the comparator makes a series of current pulses. Whenever a zero-crossing is detected by the logic system, the train of balancing pulses will be stopped. This makes the remaining voltage to stay around zero volt. Subsequently, this approach is verified through simulation with a high voltage 0.18μm technology.
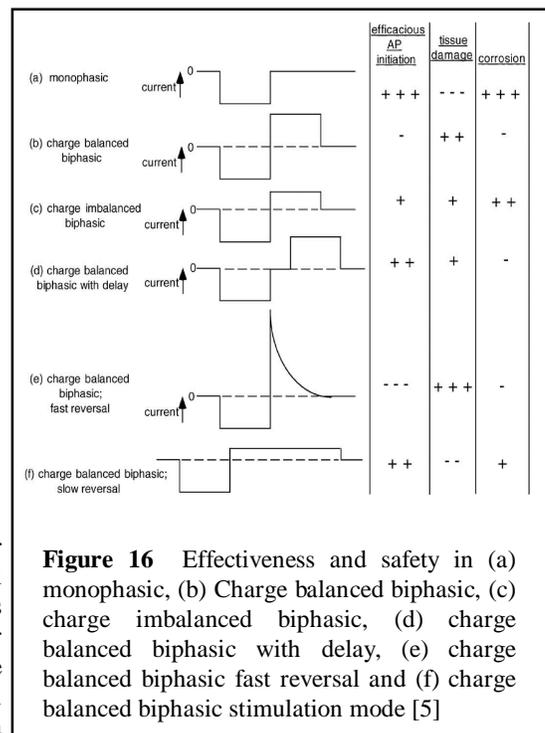
Keywords: Charge balancing, active charge balancing, functional electrical stimulation (FES), neural electrical stimulation (NES).

1. Introduction

During the past decade, the development of implantable functional electrical or neural stimulators has had a great role in treatment of neuronal or muscular disabilities such as deep brain stimulation, cochlear implant, cardiac pacemaker, and retinal implant [1]–[3]. The purpose of these stimulators is to restore the lost functions of damaged tissues. Neural stimulation can be done by injecting charge to the nerve tissues over a conducting electrode. This can be achieved by applying a constant stimulation current or voltage for a specific period of time. The most efficient stimulation mode is constant current since the amount of charge that is applied to tissue is controlled by the amplitude of the applied stimulation current [4]. However, for the constant voltage based stimulation this could be done by monitoring the amount of current which is applied to the tissue.

In order to avoid electrode dissolution or tissue destruction the stimulator system has to ensure that no residue charge remains at the electrode-electrolyte interface [5]. Hence, the stimulator must operate in two phases: cathodic and anodic. In the cathodic phase the charges are injected to the tissue to initiate an action potential, however in anodic phase the charge is drawn from the tissue (Fig. 1 b-f). For safety issues, the amount of charge in these two different phases should be equal. But, in integrated stimulators there might be more than 1%-5% mismatch between the two phases due to mismatches between stimulation amplitudes and/or pulse width. Therefore, every stimulator has a charge balancer or charge equalizer circuit to ensure the equality of the charge in anodic and cathodic phases [6]. Some approaches have been presented for charge balancing implementation. One of the basic methods is to insert a large off-chip capacitor in series with the stimulation electrode which can prevent flowing of a DC current to the tissue [7]. These capacitors have large area penalties which is unfavorable for multichannel stimulators such as retinal or cochlear stimulators. To solve these problems in a multichannel stimulator, one may use the active charge balancing method. In this method a feedback from the electrode voltage to the stimulator circuit is used to eradicate the remaining undesirable electrode voltage. Eradication can be done by various approaches, which however, are usually complicated, and consume considerable area and power [8].


Figure 16 Effectiveness and safety in (a) monophasic, (b) Charge balanced biphasic, (c) charge imbalanced biphasic, (d) charge balanced biphasic with delay, (e) charge balanced biphasic fast reversal and (f) charge balanced biphasic stimulation mode [5]

The purpose of this paper is to present an active charge balancing technique that can precisely balance the stimulation charge. The paper is organized as follows: in section 2 a simplified model of the electrode-electrolyte interface is

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introduced. A brief review on previously published passive and active charge balancing techniques is presented in section 3. In section 4, the proposed charge balancing method and its circuit are described. The simulation results are presented in section 5.

2. Electrode-Electrolyte Interface Model

As shown in Figure 2(a), the electrode-electrolyte interface can be modeled by a double layer capacitor (C_{dl}) representing the ideal charge transfer, a faradaic resistance (R_f) representing the oxidation-dioxidation reactions that occur at the electrode-electrolyte interface and a half-cell potential which depends on the type of the electrode used for stimulation. The comprehensive model between the two electrodes is illustrated in Figure 2(b). As shown, the two half-cell potentials neutralize each other and therefore they are omitted from the model between two electrodes. This model is used as the tissue impedance throughout the paper. For simulation purposes an equivalent model which is shown in Figure 2(c) is used. R_s is chosen to be 300Ω and the equivalent C_{dl} is chosen to be $1\mu F$. Due to the negligible impact of R_f, it is eliminated from the equivalent model.
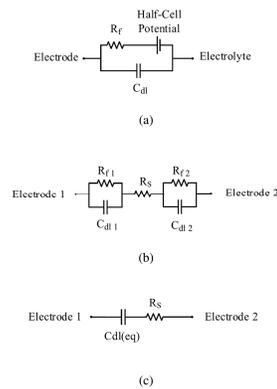

Figure 17 (a) Simplified model of an electrode-electrolyte interface and (b) model of two electrodes and the electrolyte between them. (c) Equivalent model used for simulation.

3. Literature Review

Charge balancing methods are categorized in two major groups: passive and active. In active charge balancing approaches a monitoring system is employed to monitor the remaining voltage on C_{dl} (in the absence of any current through the tissue) and equalize the charge of the cathodic and anodic phases. On the other hand, in a passive charge balancing method there is no such monitoring system and balancing is made by discharging phases. The details of these systems are investigated in the following subsections.

3.1 Passive Charge Balancing

As shown in Figure 3(a), the most common solution for charge balancing is to serialize a capacitor, namely blocking capacitor, with the electrode to accumulate the charge that flows through the tissue in cathodic phase. In the anodic phase, the blocking capacitor depletes its charge through the tissue via a switch. However, this method suffers from two major drawbacks. First, due to the limitation in the voltage headroom, the blocking capacitor must be large enough (in order of several micro Farads) and the occupied area by these off-chip blocking capacitors may induce problems in some applications such as multichannel retinal stimulation [9]. Second and the most important drawback is that the anodic duration is not controllable for different stimulation parameters and electrode impedances. As was mentioned, the blocking capacitor should be as large as possible. This increases the time constant of the discharging phase, and increases the possibility of terminating the discharging phase prematurely, reaching the stimulation phase [7].
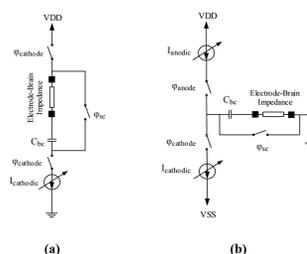
Figure 18 Two types of passive charge balancer. (a) Only cathodic pulses generated by the stimulator. (b) Both anodic and cathodic pulses generated by the stimulator.

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To resolve the latter problem, as shown in Figure 3(b), one can exert a discharging phase after the anodic and cathodic pulses, to discharge the blocking capacitor through the tissue. By means of the mentioned technique, the time required for the discharging phase is reduced significantly.

3.2 Active Charge Balancing
Eliminating the blocking capacitor allows us to introduce a monitoring system to balance the net charge of stimulation. Charge balancing systems in which a monitoring system is used, are called active charge balancers. In recent years some active charge balancing methods are proposed which are surveyed in this sub section.

One of the oldest active charge balancing method backs to a design which was introduced in [10]. In this system the anodic and cathodic currents are sampled and scaled down. This scaled current charges and discharges an integrated capacitor in order to integrate the stimulation current and thus calculate and balance the net charge.

Another method which is the basis of the proposed charge balancer is introduced in [11]. As shown in Figure 4 after each anodic and cathodic phase, the remaining voltage is compared with two levels of voltage (± 100 mV) by means of a window comparator in the absence of any current. Any remaining voltage out of this window forces the stimulator to inject current pulses to hold the remaining voltage in the pre-defined window. However, using the window comparator leads to some DC current which is not preferable for a safe stimulator. Therefore, our proposed charge balancer aims to use a single comparator to hold the remaining voltage near zero.
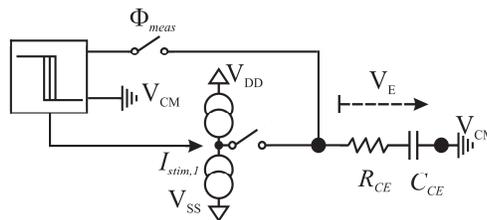
Figure 19 Pulse insertion method based on window comparator

In [12] another approach for charge balancing is used (Figure 5). After each cathodic and anodic phase, the voltage across the electrode is quantized with a window comparator. According to the output of the window comparator, a background current is continuously exerted to the tissue. However, this method suffers from stability problems as its authors mentioned.
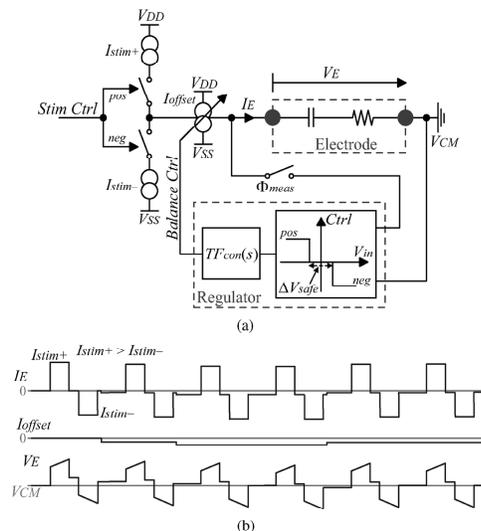

Figure 20 Background offset regulation technique [12].

In addition to the mentioned methods, another approach for charge balancing is to measure the charge by counting discreet charge packets. As described in [13], this can be done by switching the stimulation current between two capacitors and charging them to a defined level of voltage (V_{ref}) and discharging them sequentially with a shorting switch (Figure 6). Therefore each packet of charge is equal to:

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$$Q_{packet} = CV_{ref} \qquad (1)$$

Consequently, charge balancing is done by measuring the quantity of the packets in each phase.
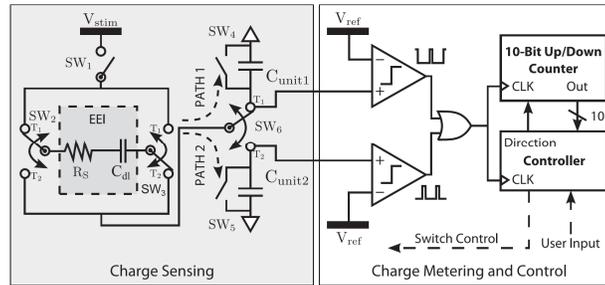
Figure 21 Charge-metering method proposed in [13].

4. Proposed Charge Balancing Technique

To achieve nearly zero remaining voltage on C_{dl}, a balancing technique is proposed in this paper. The proposed technique works by inserting some extra current pulses like in [11], but it has a major difference. As shown in Figure 4, in previous pulse insertion systems a window comparator is used to activate the charge balancing system. In a dead-zone region (which is placed between ±100 mV), the charge balancing system isn't activated since it doesn't use a zero-crossing detection mechanism. This makes the mentioned pulse insertion technique less accurate. Consequently, the safety of the system is reduced as well. This paper proposes a charge balancing system which uses a single dynamic comparator along with a logic system to detect any changes in the polarity of the remaining voltage in order to stop insertion of extra current pulses. Therefore, the remaining voltage is forced to stay close to zero. Consequently, the proposed charge balancing method is safer than its predecessors.

Figure 7 shows the proposed system for charge balancing. In absence of stimulation current, the output of the dynamic comparator shows the polarity of the remaining voltages, "0" for negative voltages and "1" for positive voltages. The goal of the proposed charge balancing system is to detect the polarity of the remaining voltage at the beginning of the charge balancing phase (ϕ_{CB}) and to inject discreet high frequency current pulses until the polarity of the remaining voltage changes. As shown in Figure 7, sign detection system adopts a simple design. In the charge balancing phase, the remaining voltage is compared with the common mode voltage and the result is saved in the comparator D-Flip Flop (CompDFF in Figure 7). According to the output of the dynamic comparator, the corresponding switch is turned on to sink or source a current in order to force the remaining voltage to change polarity. The first result of the comparison in the charge balancing phase is saved in the polarity D-Flip Flop (PolDFF in Figure 7). By using an XNOR gate, every output of the dynamic comparator is compared with the first sample of the remaining voltage which is saved in the polarity D-Flip Flop. After a change in polarity is detected, the pulse injection stops.
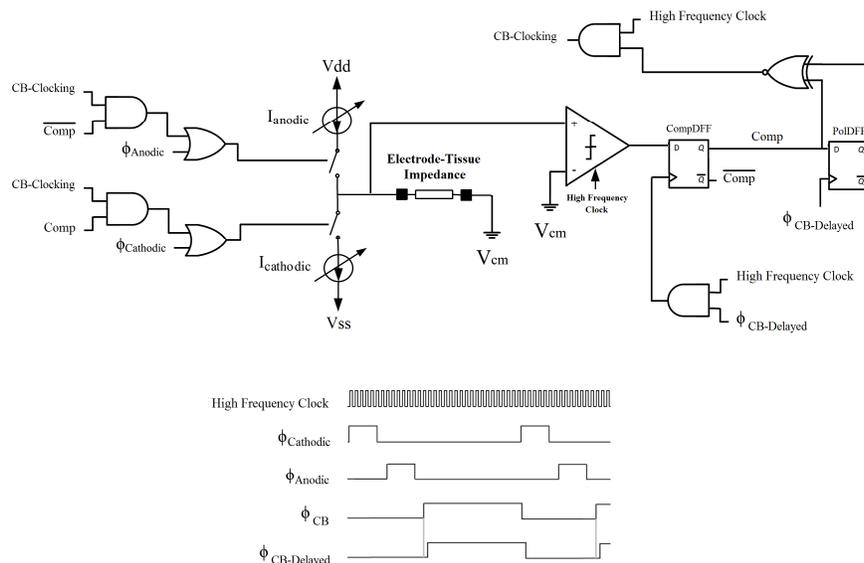

Figure 22 The proposed charge balancing system based on detection of polarity changes of the remaining voltage

5. Simulation Results

For the proof of concept, the proposed charge balancing system is simulated in high voltage 0. 18μm technology. The simulation results are shown in Figure 8. Figure 8(a) is the stimulation current with 10% mismatch between anodic and cathodic phases (the cathodic current is larger than anodic one). Figure 8(b) is the corresponding voltage of the tissue model.

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(a)

(b)
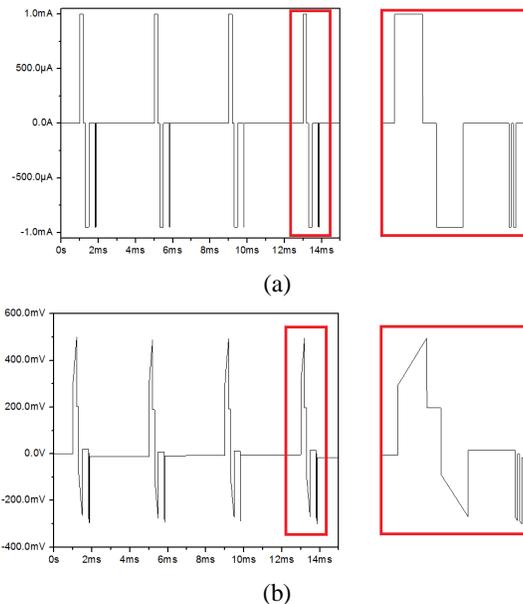
Figure 23 (a) The stimulation current with 10% mismatch between anodic and cathodic phases. (b) Corresponding voltage of the tissue model.

As it can be seen, due to larger cathodic charge, the remaining voltage before the equalization phase is higher than zero. Therefore the balancing system injects negative current pulses until it detects a change in the polarity of the remaining voltage. As a result, the remaining voltage is forced to stay near zero.

6. Conclusion

In this paper a new precise active charge balancing method is proposed. This method can be adopted in a stimulator for safe neural stimulation. By using detection of polarity changes of the remaining voltage, this method offers precise charge balancing and induces minimal power and area overhead to the circuit. This method is based on a previous pulse insertion approach, but it also utilizes zero-crossing detection in order to significantly increase the precision of charge balancing.

Simulations results are acquired in high voltage 0.18μm technology and the results for 10% mismatch between cathodic and anodic current pulses (which is more than the maximum mismatch value reported, namely 5%) show that the residue voltage left on C_{dl} when using this method is no more than ±10mv for 1mA stimulation current pulses.

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