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Recent Advances in Charge Balancing for Functional Electrical Stimulation

Kriangkrai Sooksood, Thomas Stieglitz and Maurits Ortmanns

***Abstract*— Charge balancing is a major concern in functional electrical stimulation, since any excess charge accumulation over time leads to electrolysis with electrode dissolution and tissue destruction. Its major function is to ensure that the mean value of electrode voltage is kept within a safe level. However, it serves as a failure protection as well. This paper presents an overview on recent advances in this field, both passive and active (closed-loop) charge balancing techniques.**

I. INTRODUCTION

The development of biomedical implantable functional electrical or neural stimulation (FES/FNS) has made great progress in the treatment of neural or muscular disabilities, such as cochlea implant, cardiac pacemaker and retinal implant [1], [2], [3]. The principle is to excite a neural reaction upon the transfer of charge into the tissue, either by applying a constant voltage or a constant current. However, the most efficient and popular method in biomedical implant is a constant current based stimulation. In order to avoid the irreversible electrochemical reactions such as pH change, electrode dissolution as well as tissue destruction, it has to ensure that no net charge appears at the electrode-electrolyte interface. Employing capacitor electrode is a simply and inherently safe method, but the maximum charge per unit area is less than that for noble metal electrode, e.g. platinum (Pt) [4]. Alternatively, the balanced and biphasic stimulating current pulse is typically concerned. This ensures that no net charge appears at the electrode after each stimulation cycle and the electrochemical processes are balanced to prevent net dc currents [3].

But especially when integrated circuitry is used for the stimulator, due to imperfections of the fabrication process more than 1%–5% of the mismatch of the current pulses has to be taken into account. Therefore, measures to achieve charge balancing are typically implemented. In the past different possibilities have been presented to prevent this during stimulation. The purpose of this paper is to review the most commonly adopted techniques and recent advances in this field.

II. BIPHASIC CURRENT STIMULATION PULSE

As previously mentioned, a very straightforward way to avoid the charge accumulation is the adoption of biphasic current pulses. Any charge delivered during the first phase is balanced during the second phase of the pulse. Ideally, biphasic stimulation pulses do not show net charge transfer. Due to the current mismatch, this alleviates and ends up with a net dc current and a resulting excess potential over an electrode. Measurements show that an average dc current must be kept below 10nA [5], in order to prevent a charge accumulation on the electrodes and resulting a strong faradaic current. If a stimulation current is $\pm$1mA, the biphasic current pulse matching should be in the range of 0.001% which can not be achieved in a real integrated circuit.

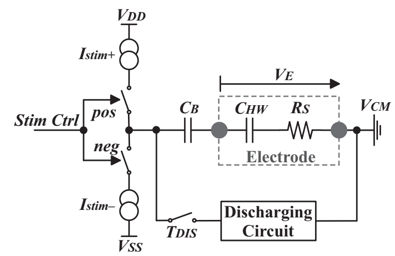

Fig. 1. Biphasic current stimulator with passive charge balancing

In order to generate a matched biphasic pulse, many techniques have been recently developed. For example, feedback DAC calibration [6], and S/H dynamic current balancing [7], which achieve lower than 0.5% current mismatch. However, an electrode shortening is still needed in such generators.

III. PASSIVE CHARGE BALANCING

A. Blocking Capacitor

To force charge balanced stimulation, the most common solution is to insert a large (several μF range), non-integrated dc blocking capacitor in series with the stimulation electrode as shown in Fig. 1. It prevents dc currents and is an effective safety feature in case of semiconductor failures [8]. Nonetheless, regular discharge of the blocking capacitor is necessary in order to avoid saturation due to dc current integration and consequently reduces the output voltage compliance of the stimulator.

This technique comes with two major disadvantages. Firstly, a blocking capacitor must integrate the total stimulation current. Thus, it is usually much larger than the electrode-electrolyte interface capacitance C_{HW} and must be externally realised. Additionally, one capacitor is needed per electrode for parallel stimulation. For a multi-channel application, such as retinal implant, this can not be implemented in the required number due to space limitation.

A promising technique to reduce the size of a blocking capacitor has been recently introduced in [8]. It is based on

K. Sooksood and M. Ortmanns are with the Institute of Microelectronics, University of Ulm, 89081 Ulm, Germany kriangkrai.sooksood@uni-ulm.de
T. Stieglitz is with the Laboratory for Biomedical Microtechnology, Department of Microsystem Engineering (IMTEK), University of Freiburg, 79110 Freiburg, Germany

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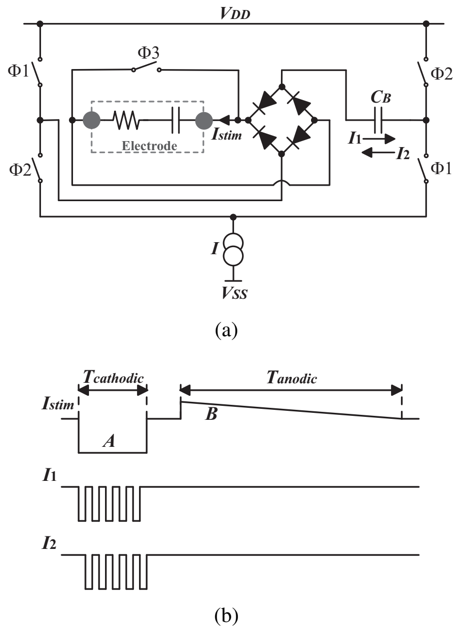

Fig. 2. (a) Stimulator with HFCS technique and (b) waveforms [8]

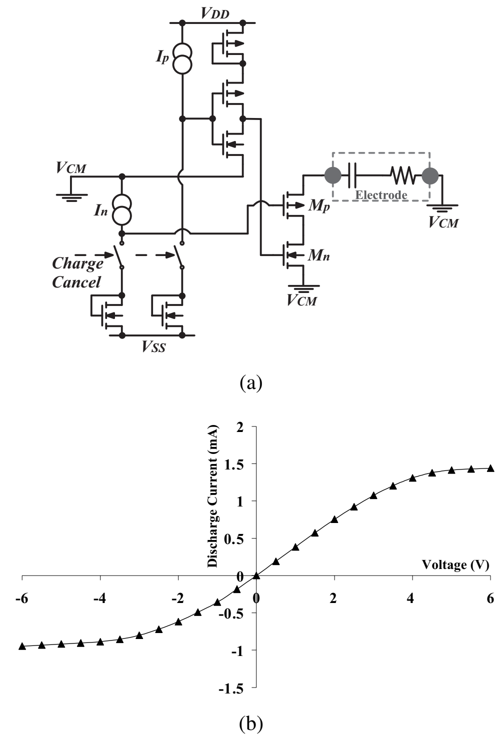

Fig. 3. Limiting current discharging; (a) circuit and (b) I-V characteristic [10]

the idea that shorter charging periods require also smaller blocking capacitors. In the illustrative circuit in Fig. 2, one phase of the stimulation current (cathodic) is generated by the summation of two high-frequency complementary current pulses, while the anodic phase is realised passively by electrode shortening through a switch Φ_3. By using a train of 50ns pulses, for example, a blocking capacitor can be as small as 100pF. This method has been improved to be a fail-safe and tested *in vitro* [9]. However, if a 100-channel stimulation is required, for example, the amount of 100-blocking capacitors still employs a very large area and can not be integrated together with the stimulator output stage. Furthermore, switching stimulation currents in the 50ns range might not even be feasible, since many stimulators need high voltage outputs, which are hardly switchable within the ns-range except with very high power consumption. This reason also prevents the further reduction of the pulses and the blocking capacitor.

Finally, the technique avoids one of the major advantages of external blocking capacitors, which is the IC failure safe operation!

B. Electrode Discharging

Because of the space limitation for blocking capacitors, recent integrated FES implementations used charge balancing relying on electrode shortening [7] or current-limited discharge circuitry at the stimulation electrodes [10], as the circuit illustrated in Fig. 1 without C_B. In a simple electrode shortening, the required shortening period depends on the time constant of the electrode and its discharging impedance. If a small size electrode is concerned, the impedance of which is very high. Then, a long shortening period is needed. To reduce required shortening time, this technique is normally used together with a precision biphasic current generator. The discharge circuitry (Fig. 3) is improved for a high voltage application in order to prevent large currents through the tissues when discharging. The limiting currents can be set by sizing the transistors M_p and M_n.

A general disadvantage of all passive discharge techniques is that their success is not controlled. After a stimulation period, a discharge process is initiated for a certain amount of time. Especially, since the current mismatch as well as the electrode impedance vary independently, the required discharge time is mostly unknown and based on experiments. Thus, the electrode potential is not known during or after the discharge period.

C. Quasi-Static Electrode Potential Expression

In the following, the derivation of an analytical expression for the quasi-static mismatch voltage on an electrode employing shortening is found. Normally, for the blocking capacitor charge balancing, the size of this capacitor must be large to reduce its voltage drop. The effect of the counter electrode is also neglected, because it normally has a much larger area than the stimulation electrode. Then, the time constant of electrode-electrolyte phase boundary simplifies to be the multiplication of the double layer interface capacitor of the stimulating electrode C_{HW} and the solution spreading resistance R_S, that is $\tau = R_S \cdot C_{HW}$. Assuming discharging period t_{DIS} is much shorter than τ, then the voltage at the stimulating electrode remains constant. An experimental study in [11] shows that the quasi-static electrode potential during non-stimulation period depends not only on the charge mismatch but also the period of discharging. This can be

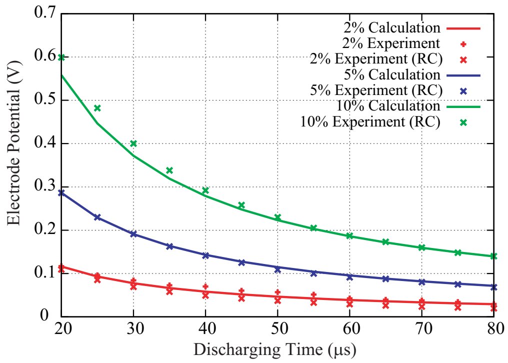

Fig. 4. Quasi-static electrode potential over discharging time on a 150μm diameter electrode

calculated from

$$V_E = \frac{I_{DC} \cdot t_{FRAME} \cdot (R_S + R_{DIS})}{t_{DIS}} \tag{1}$$

where V_E is the quasi-static electrode potential during the discharge phase. I_{DC} is the dc current resulting from the charge mismatch. t_{FRAME} and t_{DIS} are stimulation and discharging periods, respectively. Also, R_{DIS} is assumed to be the resistance of the discharge circuit in Fig 1. This equation can be used in either the case of stimulator with blocking capacitor or electrode shortening only.

The plot in Fig. 4 shows the results when a small size electrode (150μm diameter) is used with 2%, 5% and 10% current mismatches. For small mismatch (2%) the results on the electrode in saline solution match well with the calculation from Eq. 1 as well as the experimental results on the extracted RC–model. For 5% and 10% current mismatches, we observed that the electrode was destroyed during the experiment. Therefore, only the results on the RC–model can be shown here, which also fit with the calculation. These results show that if discharging time is short, the quasi-static electrode potential rapidly increases which is harmful for the electrode as well as the neural tissues.

IV. Active Charge Balancing

Basically, the resulting electrode voltage from a charge mismatch is the voltage at the capacitance part of the stimulating electrode. From the voltage-current differential equation of a capacitor $i = Cdv/dt$, if its voltage remains constant or limited, the average current over a time will be zero. To keep this voltage within a limited range, the electrode voltage will be monitored and compared with a predefined acceptable range. This process is performed during non-stimulation periods. If the residual voltage left at the stimulating electrode exceeds this range, the residual charge will be actively compensated for. This can be done by inserting a short pulse train as introduced in [1] or using an offset regulation [13], [14].

A. Pulse Insertion Technique

The simplified concept of this technique is illustrated in Fig. 5 [1]. After each stimulation pulse the switch T_{meas} will be closed shortly to monitor the electrode voltage. If this voltage exceeds a safe window, e.g. ±100mV, a predefined short current pulse with a fixed amount of charge will be generated to reduce this voltage. This routine is repeated until the electrode voltage is within the safe window. As shown by the measurement result on 5% current mismatch in Fig. 6.

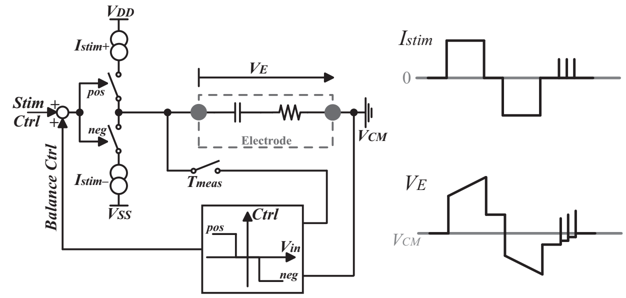

Fig. 5. Pulse insertion charge balancing and its waveforms

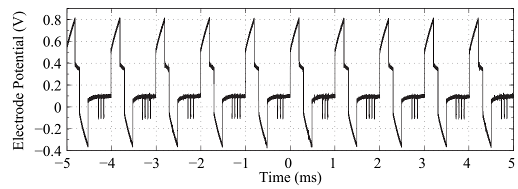

Fig. 6. Steady-state electrode potential of stimulator with pulse insertion

The technique has been successfully implemented in a HV retinal stimulator [1] and proven to be reliable. But the effect of the inserted short pulses on an unwanted neural stimulation has not been proven yet. Additionally, the required number of balancing current pulses and therewith the duration of the charge balancing depends on the actual charge imbalance after each stimulation. Vice versa, the maximum amount of mismatch charge, which can be compensated, depends on the adjusted charge per pulse and the number of pulses allowed over time.

B. Offset Regulation Technique

One possibility to avoid the balancing current spikes as well as timing conflicts with subsequent stimulations is to spread the balancing current over time. Thus, the compensating charge can be supplied in the background as an offset current. This can be done by either short monitoring after each stimulation period (Fig. 7(a)) [13] as in the previous or continuous measuring (Fig. 7(b)) [14]. If the electrode voltage exceeds a predefined window, an offset current into the electrode is adjusted for compensation (Fig. 7(c)). Hence, after an initial settling process, the background offset current cancels perfectly the mismatched biphasic stimulation current. However, for continuous monitoring, capacitive coupling is required to prevent the unwanted voltage drop during the stimulation period. This is not feasible for integration.

This offset technique is different from the before mentioned pulse insertion technique, since now the charge imbalance is not eliminated after each single stimulation, but

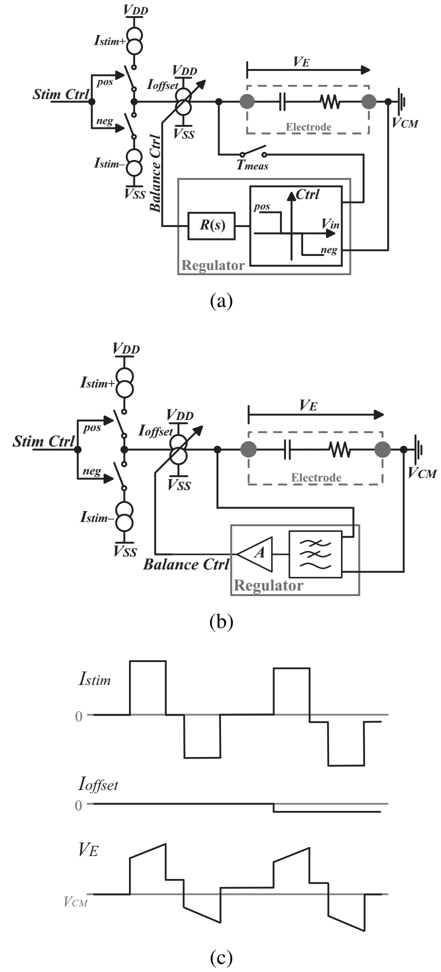

Fig. 7. Offset regulation charge balancing; (a) discrete-time monitoring [13], (b) continuous-time monitoring [14] and (c) waveforms

the charge balancing becomes a continuous background operation. After an initial settling process, the offset current is adjusted to continuously match the biphasic stimulation mismatch which keeps the electrode potential within an acceptable range. This can be observed by the measurement result on 5% current mismatch illustrated in Fig. 8.

V. CONCLUSION

This paper presents an illustrative overview on charge balancing techniques for functional electrical or neural stimulation. The most commonly adopted techniques are explained and their advantages and disadvantages are outlined. It turns out that blocking capacitor is the most effective way, if low number of stimulating sites are concerned or external components can be allowed. Its function is not only to balance the charge but also to act as a semiconductor failure protection. Together with the high-frequency current-switching technique, capacitor size can be reduced. This technique is not useful for high-frequency stimulation because it relies on a passive discharge. The switched passive discharge is an easy approach, if no real HV application is intended nor an active control of the balancing process is required. However, to reduce a discharging period when a small size

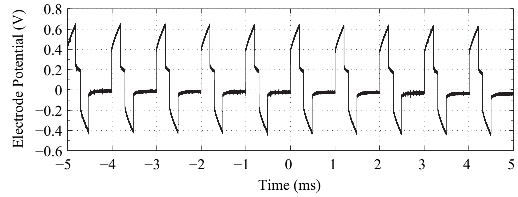

Fig. 8. Steady-state electrode potential of stimulator with offset regulation

electrode is used, a matched biphasic current generator is needed. A current-limited discharge circuit is presented to prevent high currents through the tissue during discharge in HV applications. For multi-channel implantable applications, active charge balancers are an interesting solution. They rely on the measurement of the excess electrode voltage. If stability constraints are regarded, these techniques provide operators with valuable safety information about balanced charge or safe electrode condition. They are also compatible for HV applications. Combination among passive and active charge balancers is a promising idea, to provide both a charge balancing and a fail-safe operation.

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