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A voltage-controlled current source with regulated electrode bias-voltage for safe neural stimulation

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ABSTRACT

A current source for neural stimulation is presented which converts arbitrary voltage signals to current-controlled signals while regulating the offset-voltage across the stimulation electrodes in order to keep the electrodes in an electrochemical state that allows for injecting a maximum charge. The offset-voltage can either be set to 0 V or to a bias-voltage, e.g. of a few 100 mV, as it can be advantageous for fully exploiting the charge injection capacity of iridium oxide electrodes.

1. Introduction

For the electrical stimulation of nervous tissue one of two methods of delivering a stimulus can be applied. The first one is to generate a pulse with controlled voltage amplitude, such as done in cardiac pacemakers or implantable neuromodulators for treatment of incontinence, chronic pain or symptoms addressable by deep brain stimulation. The second method, which is widely applied in the field of neuroprosthetic research and also for some clinically used devices such as cochlear implants and vagus nerve stimulators, is the generation of electrical pulses with controlled current amplitude (Prutchi and Norrid, 2005). This option has some advantages over the voltage-controlled approach: the stimulation thresholds stay more or less constant even with changing electrode impedance and ingrowth of tissue into the neural interface (Loeb et al., 1991). Furthermore, the knowledge of the current amplitude and the exact timing of the stimulus allows for the calculation of the injected charge. This is of major interest since the charge injected into an electrode of known capacitance determines the electrode voltage across its phase boundary. Electrochemical reactions are determined by this voltage. When this voltage stays within a window specific to the electrode material, electrode corrosion and hence tissue damage can be prevented (Donaldson and Donaldson, 1986a). However, the generation of current-controlled pulses is more demanding in terms of electronic hardware design.

In the past, current controlled pulses were postulated to be electrochemically safe as long as they are charge balanced, which means that charge injected during the first phase of a stimulus is entirely recovered by a following counter phase (Donaldson and Donaldson, 1986a). The charge balancing is traditionally obtained in one of two ways: (1) passively and (2) actively. Both ways provide comparable electrochemical safety for tissue and electrodes (Donaldson and Donaldson, 1986b).

In order to generate a passively charge balanced pulse, the stimulation current charges a capacitor switched in series to the stimulating electrodes, which subsequently is discharged during the counter phase (Donaldson, 1987; Haugland, 1997; Loeb et al., 1991; Smith et al., 1998). A variation of this concept involves charging a capacitor with a current equal to the stimulation current and applying a charge cancelling current phase, lasting until the charge is entirely removed from this capacitor (Gudnason et al., 1999). As long as the capacitor has a negligible leakage resistance, e.g. a bipolar capacitor with a PTFE, polyester or polypropylene dielectric (Horrowitz and Hill, 1989) these concepts are very safe and reliable. However, the charge recovery phase cannot be of arbitrary shape, but is usually fixed by the choice of circuit components, such as the discharge resistor and the capacitor itself. Fast repetition rates cannot be obtained as needed, e.g. for high-frequency alternating current neural blocking waveforms (Kilgore and Bhadra, 2004; Schuettler et al., 2004).

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Active charge balancing offers more flexibility. Commonly, stimulation pulses are generated as voltage signals, which are converted by a voltage-to-current converter to current-controlled pulses (Gwilliam and Horch, 2008). Carefully selecting the shape of the generated voltage signal allows for a charge-balanced bi- or multiphasic stimulation current. The intrinsic problem of this approach is the need for a perfectly balanced voltage waveform (which could be achieved by high-pass filtering) and an ideal linear voltage-to-current converter. The later is the major problem, since tolerances in the characteristics of electronic components introduce errors in the voltage-to-current conversion and usually compromise the charge balancing. These errors can be very small but the charge that is not recovered after each pulse adds up quickly from pulse to pulse and can drive the potential across the phase boundary of the electrode into regions beyond the water window (which is defined as the electrode material-specific voltage range within which charge transfer does not cause hydrolysis of the electrolyte) or at least compromises the amount of charge that can safely be injected per pulse without causing electrode corrosion.

The work presented here can be interpreted as a combination of the advantages of current controlled stimulation, i.e. stable stimulation thresholds and known charge per phase, based on a voltage-to-current converter (hence: freedom of shaping the pulse) and voltage stimulation, which sets the offset-voltage across the electrodes to zero (or any other offset-potential) and therefore avoids a drift of the electrode potential to electrochemically risky regions.

2. Materials & methods

2.1. Concept

The general concept of the circuit is simple: a commonly used voltage-to-current converter for grounded loads, the *Howland Current Source* (Horrowitz and Hill, 1989) also named *Howland Current Pump* (Prutchi and Norrid, 2005), is used for converting a voltage waveform into a current-controlled waveform, which is led through the electrodes into the tissue. The voltage across electrodes and tissue is observed and, after buffering and low-pass filtering, mathematically subtracted from the input voltage signal. As soon as an offset-voltage potential builds up across the electrodes over time, the voltage-to-current converter adds or subtracts current in order to bring the electrode voltage back to zero volt, or any other pre-set voltage.

2.2. Circuitry

The input voltage signal V_{IN}, generated by an external source, is fed into the subtractor circuit (Fig. 1), whose output voltage is converted to a current signal i_{OUT} flowing through the electrodes and

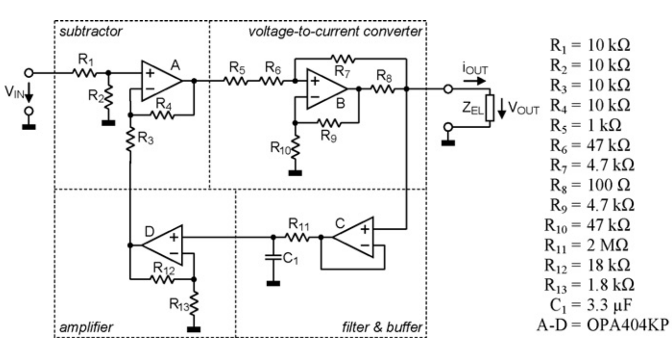

Fig. 1. Schematic and component list of the stimulator circuit.

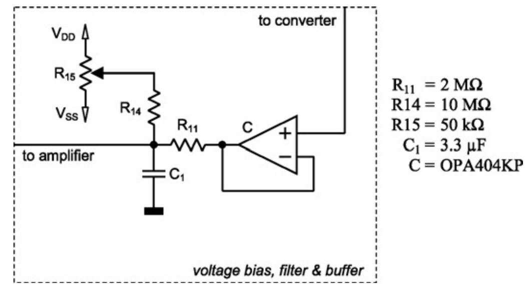

Fig. 2. Schematic and component list of the functional block: *voltage bias, filter & buffer*.

tissue Z_{EL} into the electronic ground. The voltage V_{OUT} across Z_{EL} is buffered and low-pass filtered. After 10-fold amplification, this signal is subtracted from the input voltage V_{IN}, causing the current generated by the voltage–current converter to change accordingly. The feedback loop via buffer, filter, amplifier, and subtractor is set in a way that the offset-component of V_{OUT} is always regulated to zero volts. The component values as proposed in Fig. 1 lead to a voltage to current ratio of 500 Ω, e.g. $V_{IN} = 1$ V is converted to $i_{OUT} = 2$ mA. The −3 dB cut-off frequency of the low-pass filter is set to $f_C = 1/(2\pi R_{11} C_1) = 24$ mHz. The four operational amplifiers *A–D* are one single integrated circuit (IC): the quad operational amplifier OPA404KP by Burr-Brown, available from Texas Instruments (Dallas, Texas, USA). The operational amplifiers define the maximum supply voltage of the circuit, which in this case is $V_{DD} = -V_{SS} = 18$ V, as well as the maximum output current i_{OUT} at low impedance loads, here: ±10 mA, and the output voltage compliance (±13 V).

In order to keep the schematic clear and easy to understand, the voltage supply of the amplifiers is not drawn in Fig. 1. For a sufficient stability of the circuit, two 1 µF capacitors were connected between ground and each supply rail (V_{DD} and V_{SS}), located in close proximity to the amplifier IC.

When iridium oxide electrodes are used for stimulation, an electrode offset-voltage different to 0 V is advantageous in order to make use of the large charge injection capacity of this specific electrode material (Cogan et al., 2006). Such voltage biasing can easily be integrated into the proposed circuit by exchanging the functional block *filter & buffer* (Fig. 1) by the block *voltage bias, filter & buffer* (Fig. 2). The potentiometer R_{15} sets the bias-voltage across the electrode.

2.3. Evaluation

The functionality of the circuit was evaluated by applying different input signals V_{IN} to the circuit (supplied with ±15 V) while measuring the output current I_{OUT} as well as the electrode voltage V_{OUT}. All tests were performed *in vitro* using two platinum electrodes, 750 µm in diameter, immersed in phosphate buffered saline solution (pH 7.5) at room temperature. A resistor $R_M = 100\ \Omega$ was placed in series to the electrodes to measure the voltage V_M and calculate the actual current $i_{OUT} = V_M/R_M$ flowing through the electrodes.

The first test was carried out in order to evaluate the accuracy of the voltage-to-current conversion without voltage biasing with special respect to long pulse shapes. A voltage waveform V_{IN} produced by a custom build signal generator similar to that described in Gwilliam and Horch (2008) was applied to the circuit. The waveform consisted of a pulse of +250 mV amplitude and 250 µs width, followed by a −25 mV, 2.5 ms counter pulse. The pulse repetition frequency was set to 22 Hz. V_{OUT}, I_{OUT}, and V_M were recorded and digitized at a sampling rate of 200 kS/s using a digital oscilloscope (type 54622D, Agilent, Santa Clara, CA, USA).