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Technical note

Battery powered neuromuscular stimulator circuit for use during simultaneous recording of myoelectric signals

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ABSTRACT

Surface Functional Electrical Stimulation (FES) requires high stimulation voltages. A step-up transformer in the output stage of the stimulation circuit is often used. In the present technical paper a voltage controlled current source (VCCS) is presented as an alternative to the transformer coupling. Two (master–slave) coupled transconductance amplifiers (TAs)—in series with pre-charged capacitors—are used to drive the output current. After each stimulation pulse the capacitors are recharged to a high voltage by a switch mode power supply (SMPS). A multiplexer in the output stage is used to provide biphasic output. Output rise-time (10–90%) was less than 2 μs at 100 mA output. Biphasic charge balanced stimulation current can be produced with a net current to ground of less than 20 nA, thus virtually separated from ground. The circuit permits recording of the volitional myoelectric signal from the stimulated muscle. It is part of a portable myoelectrically controlled FES system powered by 2 AA batteries and currently used in clinical trials.

1. Introduction

Functional Electrical Stimulation (FES) can be used to evoke and control muscle activity in subjects with central nervous system disorders [1]. The stimulation circuit has to be small, light and energy efficient for portable FES systems and when measuring bio-potentials (such as the myoelectric signal) during stimulation, special constraints to the output current apply.

The muscle contraction can be controlled by the current level, which for transcutaneous FES typically ranges from 10 to 100 mA [1]. The electrode impedance can be 1–5 kΩ [9], so battery-powered systems are typically using a step up transformer in the output stage to drive the current [2,4,10,13]. To transfer the energy of a 300 μs stimulation pulse, the transformer becomes bulky and heavy—typically 20 mm × 20 mm × 20 mm (e.g. Elpha II 3000 by Danmeter A/S, Odense, Denmark or ODFS by Odstock Medical Limited, Salisbury, UK). Transients, caused by droop or inductance and parasitic capacitance of the transformer, may disturb recording of bio-potentials from the same tissue.

Alternatively to transformer coupling, a DC–DC converter can be used to generate the voltage needed to drive the current [3,10–12,19]. Switch mode power supplies (SMPS) can be small (6 mm × 9 mm board space) and efficient (80%) [5]. High voltage amplifiers (PA85) can be used [11], but quiescent power consumption is high—3 W according to the datasheet. Since the stimulation duty-cycle is only about 1% of the stimulation period quiescent power will drastically reduce the overall efficiency. Matched current mirrors can be a solution [19], but imbalance in push-pull stages will cause current leakage to ground and power is lost in the mirror current.

A charge balanced biphasic stimulation waveform can be composed of two rectangular pulses with opposite polarity having an intra-pulse interval (IPI) between the two phases [8]. A pulse duration and IPI of 300 μs produces a relatively comfortable and efficient contraction, using a repetition rate of 10–50 pulses per second (pps) [1,8]. A fast rise-time (<2 μs) is commonly used as a design criterion [11,19], although there are no published data on actual physiological requirements for this parameter.

Myoelectric signals (MES) can be used for controlling FES [15,17]. The recorded MES will be disturbed by stimulation artifacts (SA), when the stimulated and the controlling muscles are closely spaced or are homologue. The SA is caused by spillover from the stimulator to the amplifier electrodes. Charge balanced stimulation will reduce SA and a common mode current in the recording electrodes should be avoided. Hence, the stimulator should have high impedance towards the MES amplifier circuit [6,15].

This work presents a voltage controlled current source (VCCS) designed for: charge balanced biphasic current output, high output-to-ground impedance, minimal quiescent power consumption and double fault protection against DC current. Further weight-size optimisation is possible.

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2. Materials and methods

2.1. Specifications

The VCCS described is a sub-circuit of a portable system. The system comprises a microprocessor, a MES amplifier [16] and standard DC-converters generating ± 3 V. The system also has a MC34063-based adjustable SMPS utilizing a hand wound transformer (EP7-3E3 core). The SMPS occupies $3\,\text{cm} \times 1.5\,\text{cm}$ board space. It can deliver up to 154 mW with an output voltage from ± 77 to ± 100 V with an efficiency of 37%. The whole system is powered by 2 AA batteries.

The maximal output current (I_e) of the VCCS is chosen to be 100 mA at 1 kΩ load. A charge balanced waveform was chosen with a first (positive) phase ($t_p = 300\,\mu s$), an intra-pulse interval ($t_i = 300\,\mu s$) and a second (negative) phase ($t_n = 300\,\mu s$). The repetition rate was set to $t_r = 60$ ms.

2.2. Circuit description

The key principle is having two transconductance amplifiers (TAs) in series with a sensing resistor (R_s), where R_s is on the low voltage side (Fig. 1). Each TA is embodied as a field effect transistor (FET) where the gate is driven by an operational amplifier (op-amp). The source terminal of the FET is fed back to the inverting input of the op-amp. For TA2 the source will be virtual ground ($V_2 = 0$) and TA1 will ensure that voltage drop over R_s is equal to the positive analog input signal ($V_1 = V_i$). Hence the current in R_s will be

$$I_e = V_1 = \frac{V_i}{R_s}, \qquad (1)$$

which will be the actual stimulation current. Both TAs will have identical drain currents (I_e), since both FET gate currents and op-amp input currents are virtually zero. Being current generators the output to ground impedance is theoretically infinite, only limited by FET leakage currents, capacitive gate-source and gate-drain coupling and op-amp input specifications. The energy for draining and sourcing current, must be stored on two pre-charged capacitors C_1, C_2 and the circuit works in two stages, a charging interval and a stimulation interval, see Fig. 1.

The VCCS is having a positive (Pp) and a negative (Pn) output and biphasic stimulation is obtained by using an analog switch (multiplexer) to reverse the polarity of the second phase. Thus the same VCCS is producing both phases.

Stimulation energy is transferred from an adjustable switch mode power supply (SMPS) generating a high voltage ($\pm V_{hv}$) to the capacitors (C_1, C_2), by closing two switches (Q3, Q4). In this work the switch network is realized in a simple way using high voltage BJT transistors (type TA42, TA92) such that they can be controlled by 3 V logic. The charge interval can be 9 ms or less, depending on SMPS and transistor power ratings. Control and feedback circuits are supplied by ± 3 V to minimize quiescent power consumption. The circuit diagram is shown in Fig. 2.

A multiplexer can be realized with discrete components using FETs with a latching capacitor from gate to source. On the rising edge of the current pulse the FET can be individually latched using two logic inputs (*pc*, *nc*) to select output polarity (see Fig. 3).

The capacitors must be sufficiently charged to keep the drain-source voltage of the FETs in the active region; fulfilling the equations during the duration of each phase:

$$V_{c1}(t) > V_{ds1,on} + \tfrac{1}{2} I_e Z_e + I_e R_s \qquad (2)$$

$$V_{c2}(t) > -V_{ds2,on} + \tfrac{1}{2} I_e Z_e \qquad (3)$$

where V_{c1} and V_{c2} are the voltage drops over the capacitors, $V_{ds1,on}$ and $V_{ds2,on}$ are the drain-source on-voltage over the transistors Q_1 and Q_2 respectively and Z_e is the electrode impedance. The capacitors are charged to the same voltage $V_c = V_{c1} = V_{c2}$. Let the first phase start at the time t_0 where the capacitors are fully recharged to $V_c(t_0)$. If they are not recharged during the intra-pulse interval t_i it will follow that V_c reaches a minimum at $t = t_0 + t_p + t_i + t_n$ (i.e. the pulse width) after delivering I_e for the time $t_p + t_n$ (i.e. the duration of two phases).

After each biphasic stimulation pulse capacitors are recharged by connecting the drain-terminals to $+V_{hv}$ and $-V_{hv}$ respectively and grounding through the 'freewheeling' diodes D_1 and D_2. The capacitor voltages at the time t_0, where the positive phase starts, is $V_c(t_0) \approx V_{hv}$. The FETs (Q_1 and Q_2) must be turned off when the capacitors are recharged by setting $V_1 = V_i = 0$ which can be ensured by an offset V_{off2} on TA2.

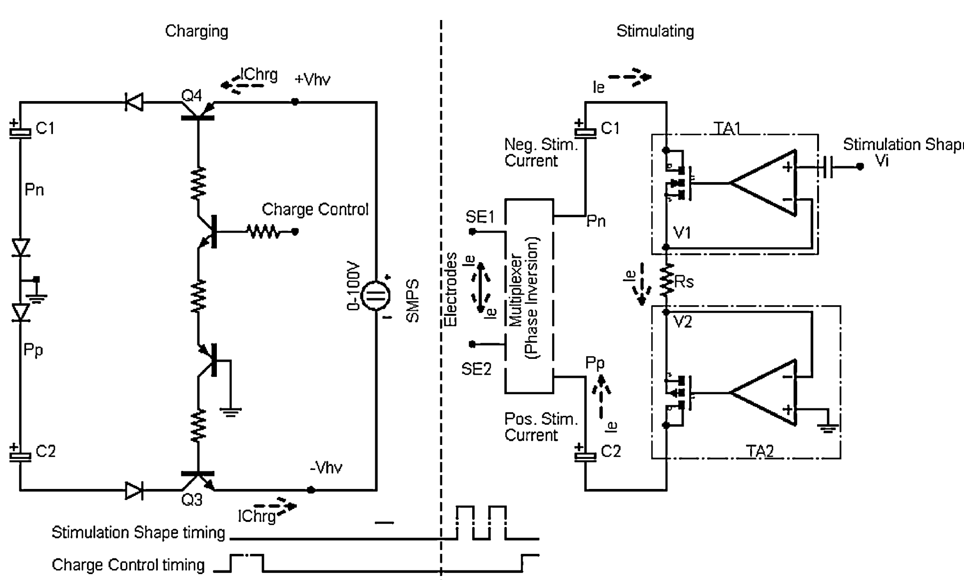

Fig. 1. Equivalent diagram of the VCCS in the two stages: charging the capacitors (left of the dashed line) and producing the stimulation current (right). A master TA_1 is driving the stimulation current and the slave TA_2 is mirroring the current so V_2 is held zero. The voltage drop $V_1 - V_2 = I_e \cdot R_s$ is held equal to $V_1 = V_i$ by TA_1 and TA_2 (patent pending MI07A00595).

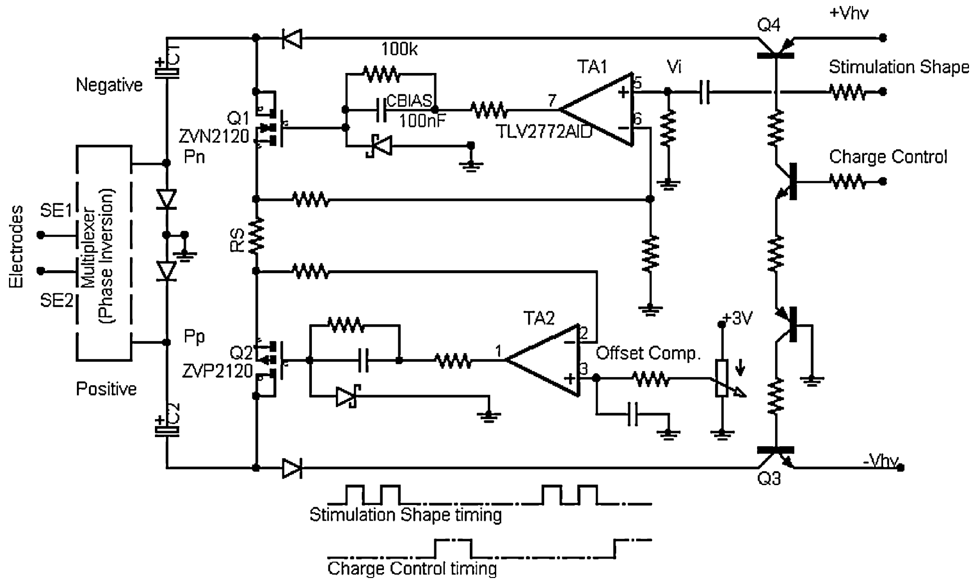

Fig. 2. Complete diagram of the VCCS. The absolute value of the stimulation pulse, the shape, must be fed to the $V_1 = V_i$ input while the polarity is controlled by using a multiplexer for phase inversion in the front end. Capacitors C_1 and C_2 are charged by from a SMPS when the *charge control* is high. To ensure pinch off of the FETs, the output of the ±3 V powered op-amps is shifted 3 V by a charged capacitor (C_{bias}). The full possible range of the gate potential becomes 0 to 6 V respectively −6 to 0 V for the Q2. The timing of stimulation and charging is shown beneath the figure.

If all multiplexer switches are closed before enabling the *charge control* signal, electrodes will be clamped to ground, through the freewheeling diodes. In the intra-pulse interval the multiplexer is unlatched. To generate charge balanced output the amplitude and duration of $V_1 = V_i$ must be the same for both phases.

2.3. Calculation of capacitors and component selection

By combining Eqs. (2) and (3) with the assumption that the capacitors are equal and identically charged, the following condition must be true during the stimulation pulse to ensure that the

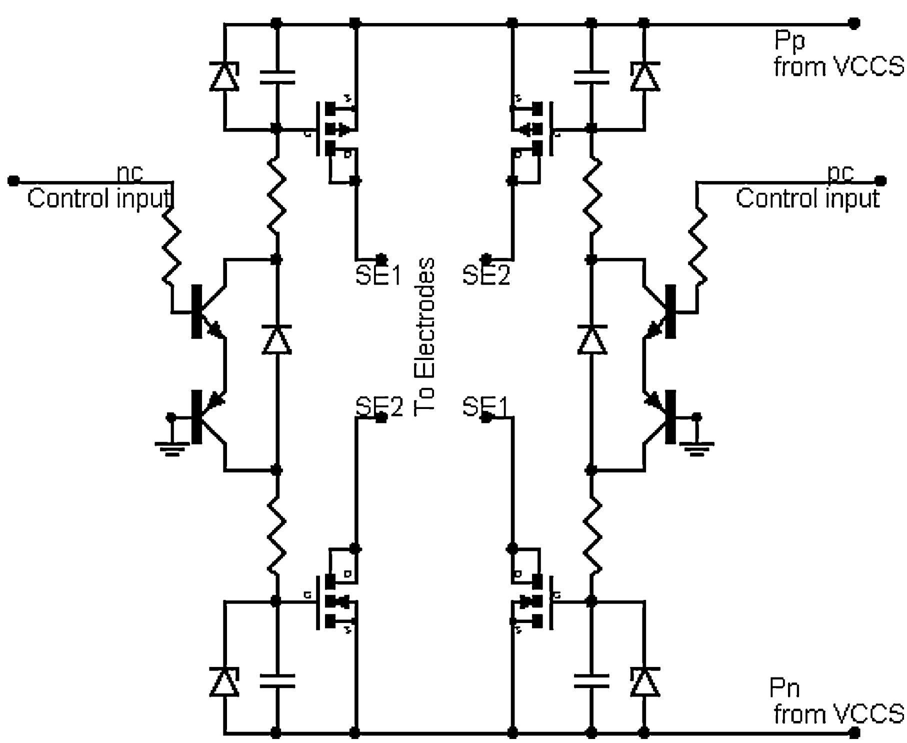

Fig. 3. The multiplexer circuit uses FETs that are latched by charging the capacitors on the rising flank of the stimulation pulse. Latching is controlled by logic signals *pc* and *nc* where *nc* will invert polarization of the current pulse. Zener diodes are protecting the gates from overvoltage. When stimulation current falls to zero the capacitors will be discharged through the freewheeling diodes of the VCCS, thus turning the FETs off for the following pulse.

FETs are in the active region (please refer to the diagram in Fig. 1).

$$V_{C1}(t) + V_{C2}(t) > V_{\mathrm{ds1,on}} - V_{\mathrm{ds2,on}} + I_e(t)Z_e + I_e(t)R_s \tag{4}$$

The voltage change (ΔV_c) for each capacitor ($C = C_1 = C_2$) when discharged by the stimulation pulses (t_p and t_n), provided that I_e is constant for both phases, is Eq. (5):

$$\Delta V_c = (t_p + t_n)\frac{I_e}{C} \tag{5}$$

The capacitor thus needs to be charged to a level given by Eqs. (2) and (3) plus this voltage drop.

The stimulator circuit was realized using high voltage FETs (Q_1 = ZVN2120, Q_2 = ZVP2120 having $|V_{\mathrm{ds(on)}}| < 1$ V at 100 mA) and high-slew-rate rail-to-rail amplifiers (TLV2772AID). A sense resistor of $R_s = 10\,\Omega$ is selected to provide a transconductance of 100 mA/V.

Selecting a capacitor size to $C_1 = C_2 = 4.7\,\mu$F the voltage drop, according to Eq. (5) becomes

$$\Delta V_c = (600\,\mu\mathrm{s}) \cdot \frac{100\,\mathrm{mA}}{4.7\,\mu\mathrm{F}} = 13\ \mathrm{V}. \tag{6}$$

According to Eq. (2) this requires:

$$V_{hv} > \Delta V_c + \left(V_{ds1} + \tfrac{1}{2} I_e Z_e + I_e R_s\right) = 13\ \mathrm{V} + 1\,\mathrm{V} + \tfrac{1}{2} \cdot 100\,\mathrm{mA} \cdot 1\,\mathrm{k\Omega} + 100\,\mathrm{mA} \cdot 10\Omega = 65\ \mathrm{V}. \tag{7}$$

2.4. Output characteristics

A resistance ($R_e = 1\,\mathrm{k\Omega}$) was used to evaluate output characteristics. To test leakage current to ground the following arrangement was made. From the midpoint of R_e a resistance ($R_{ref} = 100\,\mathrm{k\Omega}$) was connected to ground. An imbalance in the current output will result in a current to ground, causing a potential across R_{ref}. A capacitor (C_{ref} = 100 nF) was inserted parallel with R_{ref} to low-pass filter eventual current spikes. These values were chosen high to emphasize the effect of leakage currents to ground.

To test the effect of the stimulator two electrodes 5 cm × 9 cm (Pals™) were placed on the calf muscle (of the first author) accordingly to Baker et al. (p. 183) [1] with the plantar flexion isometrically held at 90°. The voltage over the electrodes was measured stimulating at the highest tolerable current (80 mA) for 10 s. Myoelectric signal from the same muscle was recorded by 3 ECG electrodes (Kendall ARBO, Ø 24 mm) in a bipolar recording configuration with the ground electrode in the middle. ECG electrodes were placed on a center line perpendicular to the stimulation electrodes to reduce the origin of stimulation artifacts [6,15]. Charging and electrode clamping was activated after 1.9 ms. A specialized amplifier with a gain of (74 dB) and designed for SA suppression [16], was used to record the myoelectric signal from the stimulated muscle during volitional contraction. The M-wave, which is the synchronized response to the stimulus of the muscle, and residual stimulation artifacts can be considered quasi-stationary, whereas the volitional myoelectric signal is stochastic [7]. The first 10 ms can be blanked since the amplifier is saturated by the SA and the following comb FIR-filter has shown useful for estimating the volitional MES [15].

$$y(t) = x(t) - x(t - t_r) \tag{8}$$

Here $x(t)$ is the recorded signal at the time t, $y(t)$ is the filtered signal, and t_r the stimulation interval using a sampling interval of 0.4 ms.

3. Results

For the VCCS part, quiescent power consumption was 13 mW, due to the control circuit (the op-amps) and independent of stimulation output. Due to high ripple (14 V) of the SMPS output during charging of the capacitors it was not possible to accurately measure the power loss in the high voltage circuit of the VCCS part. An ampere meter with 470 μF capacitor in parallel showed 1 mA, which gives 130 mW consumption of the VCCS circuit when delivering 100 mW yielding a total efficiency of the VCCS part to 70%. The drive circuit for the switches Q3, Q4 accounts for 12 mW. The overall efficiency of the circuit is the product of the VCCS and the SMPS efficiency.

Load-midpoint voltage was measured to be below the noise-floor of the oscilloscope (5 mV) at both 100 kΩ and 100 kΩ||100 nF midpoint to ground impedance. Hence the leakage current from the stimulator output-to-ground is less than 5 mV/100 kΩ = 0.02 μA.

The 100 mA rise-time$_{(10–90\%)}$ was measured to be less than 2 μs, see Fig. 4. On the same figure the differential voltage over the electrodes when stimulating human muscle at 80 mA (maximal tolerable level of the subject). Electrode differential voltage reaches 170 V yielding an electrode resistance of 2.13 kΩ. In this case the SMPS was increased to ±100 V to avoid saturation of the VCCS. As also discussed by Perkins [9] there is a polarization potential [18] causing a transient seen in Fig. 4. The step in the transient shows where the recharging of the capacitors begins. It demonstrates that the electrodes are effectively clamped together and discharging the polarization potential.

The myoelectric signal of the volitional contraction during stimulation, contains M-wave and stimulation artifacts that saturate the amplifier for more than 20 ms after stimulation, see Fig. 5. Since the electrodes are placed perpendicular to the stimulation current the effect is minimized. The large gain required for a MES amplifier (74 dB) together with the mix of stimulation artifacts and the M-wave will cause a complex, but repeatable waveform. Fig. 6 shows a window of the digitally filtered signal starting 10 ms after stimulation onset and ending 60 ms after (where a new stimulation pulse is issued). Since the transient is a stationary signal, it will be eliminated by the filter and the output (y) is zero until the myoelectric signal becomes visible.

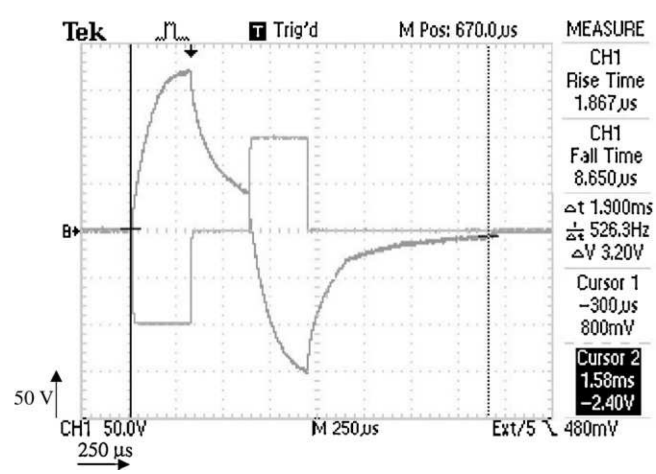

Fig. 4. Output voltages for: purely resistive load (rectangular, inversed polarity) and over a pair of 5 cm × 9 cm electrodes attached over the calf muscle. A 100 mA in 1 kΩ load shows a current rise-time (10–90%) of less than 2 μs. Electrode voltage when stimulating at 80 mA displays the transient due to the charge residue on the electrode while the stimulator output is left in high-impedance (discharging through the tissue). At 1.9 ms after stimulation onset (second vertical cursor), the effect of the output clamping on the transient can be seen. Current pulse with is 300 μs for both phases. For the electrode load the differential voltage of the first phase reaches 170 V whereas the second phase reaches only a 150 V peak value. Before the second phase the potential is approximately 40 V.

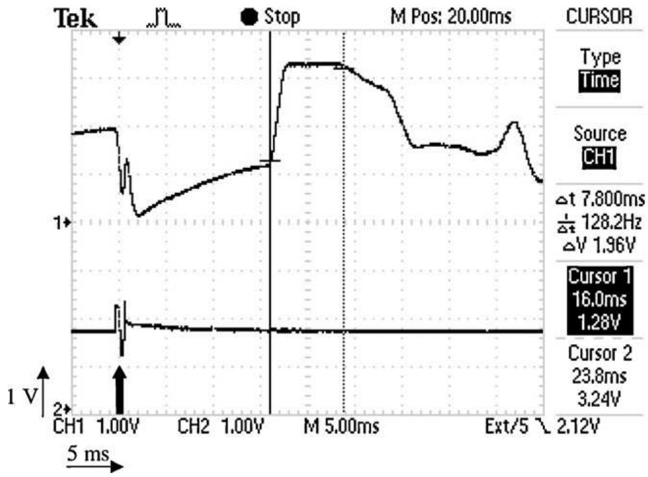

Fig. 5. The signal from the recording electrodes during stimulation and simultaneous volitional contraction. The stimulation pulse is visible at 5 ms (indicated by an arrow). Lower trace: output of preamplifier with 20 dB gain. Upper trace: raw signal from the MES amplifier with a total gain of 74 dB. For 23 ms after stimulation (second vertical cursor) the amplifier is saturated by transients (stimulation artifacts mixed with compound action potentials). Hereafter the stochastic MES is visible. The polarity of the SA is depending on the exact position of the recording electrodes with respect to the stimulation electrodes; shifting the electrodes a few millimeters can reverse the polarity.

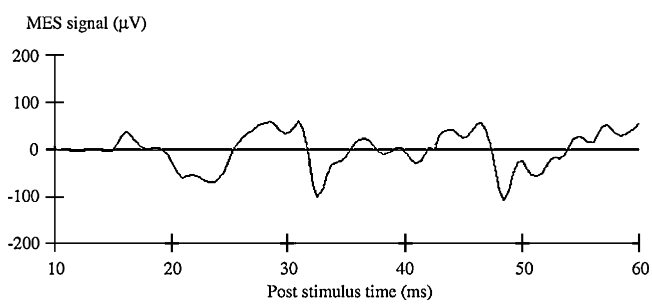

Fig. 6. Filtered myoelectric signal from volitional contraction while stimulating the calf muscle (sampling frequency 2.5 kHz). Initially the signal is flat due to the filtering of the stationary part of the transient from the saturated amplifier circuits.

4. Discussion

Optimal efficiency requires an accurate prediction of the electrode impedance charging of the capacitors. The most important point for direct synthesis of current from a high voltage source is that the quiescent power is kept low because FES applications often have an intermittent nature with periods of no output. Quiescent power consumption of the proposed circuit is limited to the dissipation of the low voltage control circuits (13 mW). Efficiency and SA related properties of the stimulators have not been documented in the literature [1–13]. In the current system the SMPS should be optimized to reach 80% efficiency, thus the overall efficiency would become 56%.

In case of insufficiently charged capacitors, the op-amps will drive the output into saturation because the absolute impedance $|Z_e|$ is higher than V_{hv}/I_e. Such detection has the advantage of not connecting directly to the high voltage side. The detection can be used for indicating a faulty condition where the electrode impedance is excessively high e.g. an electrode is falling off.

Output terminals can be clamped to ground by closing all switches in the multiplexer during recharging of the capacitors and hence actively discharge the stimulation electrodes.

The repetition rate of the stimulator is limited by the charging time of the capacitors, thus rates of more than 100 pps are possible. Duplicating the multiplexer can provide additional time multiplexed outputs thus upgrade the circuit to a multi-channel stimulator. If output power is increased by repetition rate or multiplexing requirements, the SMPS has to be chosen accordingly. Since the VCCS is an analog transconductance amplifier it can deliver any biphasic, non-continuous signal.

DC current protection is guaranteed by four components: charge switch (Q3, Q4), charge capacitors (C1, C2), freewheeling diodes and the output multiplexer. The subject is therefore protected against any single component failure.

5. Conclusion

A low-side controlled voltage controlled current source, using switched capacitors and multiplexing of the output is proposed as a neuromuscular stimulation circuit. It can provide a charge balanced biphasic current output with a low quiescent power consumption and very low leakage current to ground. It is designed for surface stimulation of muscles from which residual volitional myoelectric signal is simultaneously recorded. The circuit is part of a myoelectrically controlled stimulator, currently being tested in a clinical trial on subjects with sustained tetraplegia due to spinal cord injury [14].

Conflicts of interest

The authors declare that they have no competing interests.

Acknowledgement

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