A Universal Functional Electrical Stimulator Based on Merged Flyback-SC Circuit

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***Abstract* — This paper introduces novel functional electrical stimulation (FES) system architecture. Compared to the existing state of the art solutions [1]-[15], the presented FES system drastically increases variety of stimulation pulses, results in a less painful therapy and has significantly lower power consumption. Furthermore, the FES system inherently provides zero net charge of stimulated tissue eliminating the need for a dedicated discharging circuit. The key element of the FES is a novel power stage merging a flyback and switch-capacitor (SC) converters. The flyback steps up battery voltage and provides galvanic isolation. The following SC stage produces zero net charge high slew-rate pulses reducing pain sensation. The control of the output current pulses is performed through integrated digital voltage-programmed current mode control.**

The new FES system was tested with able bodied individuals. The results show that it produces pulses with a 10 ns rise time, which compared to other known solutions, result in the same muscle force output for 30+% less stimulation energy. The results also show that the presented FES system requires 60% less energy compared to other known systems allowing longer battery life in portable applications.

***Index Terms* — FES, DC-DC converters, digital control.**

I. Introduction

Functional electrical stimulation (FES) [1]-[17] is a technique that injects electrical pulses into an excitable tissue, i.e. nerves or muscles, to restore lost motor activity. Today, functional electrical stimulators are frequently used in spinal cord injury and stroke therapies [3]-[15] helping these individuals recover mobility, bowel and bladder function, and respiratory function. The FES is also used for pain suppression, muscle strengthening and in pre-competition warm up. Other applications include foot drop stimulation and deep brain stimulation for treating symptoms of the diseases such as pain and Parkinsonism.

Based on the types of pulses they produce, FES devices can be divided in two categories: voltage and current controlled. The voltage controlled stimulators [8]-[9] are simple to implement but are more susceptible to tissue conductance variations. The output current depends on the tissue impedance, which fluctuates within an individual patient and can significantly change during a single therapeutic session, causing problems in controlling the muscle contractions. For that reason, more complex current stimulators [4]-[7], [10]-[12] providing predictable and repeatable excitation are preferably used.

The most common shapes of current pulses used in FES therapies are shown in Fig.1. The parameters that define the pulses are: current amplitude (*CA*), pulse duration (*PD*), pulse frequency (f_{pulse}), polarity, symmetry, rise time (t_{rise}), and stimulation duration.

The pulse parameters widely vary with applications. They are also strongly related with the quality of stimulation, i.e. pain sensation. For example, the stimulation of big and/or deep muscles requires higher *CA* and longer *PW* than those for small muscle. The pulse rise time, i.e. slew rate, is related to the pain sensation. A study in [18] shows that as t_{rise} decreases the *CA* required to produce muscle contraction decreases as well. Reducing t_{rise} has two positive effects: 1) lower patient stress and pain levels; and 2) extended battery life in portable applications. Lower patient stress during therapy allows longer, more intensive sessions resulting in potentially faster recovery.

Another extremely important pulse quality measure is net charge, related to patient safety. In order to prevent excessive charge buildup in a human tissue, causing cell damage and pain, the net injected charge must be strictly controlled. For monophasic pulses of Fig.1.(a) the charge build up is limited by using very short pulse sequences and giving cells a long time to recover after that. Hence, the application of monophasic pulses is severely limited. The biphasic pulses of Fig. 1.(b) drastically reduce charge build up and associated galvanic processes that could occur in the tissues [17]. The positive portion of the pulse excites the tissue and the negative portion is used for discharge. Ideally, the pulses are created such that areas *A*1 and *A*2 are equal resulting in zero net charge.

Existing FES devices are usually optimized for a specific application [3]-[14] and/or have fairly limited

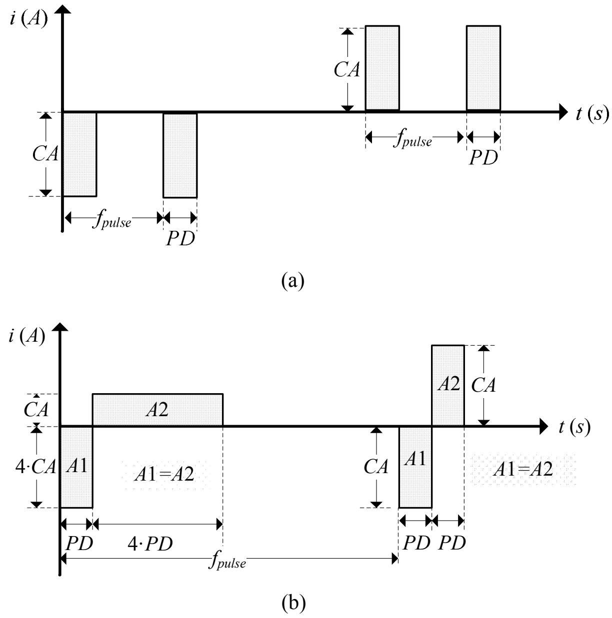

Fig. 1. Pulse characteristics for FES applications. (a) Monophasic negative (left) and monophasic positive (right). (b) Biphasic asymmetric (left) and biphasic symmetric (right).

usage time, both in terms of the therapy session duration and battery operating time. Ideal system should be able to deliver the therapy continuously during the entire day.

The limitations of existing FES solutions are mostly caused by three specific problems related to their power stages. First, the power stages are predominantly controlled by analog solutions, not offering sufficient flexibility for producing a wide variety of pulses. The second problem is caused by imperfect operation of the charge balancing circuit [19] limiting the maximum frequency of the simulations pulses [5]-[6], [10] and the third problem is caused by a fairly limited current slew rate [4]-[12] causing pain related constraints.

The main goal of this paper is to introduce a new FES architecture shown in Fig.2 that can be used for a wide range of applications and results in lower pain sensation than most existing solutions. In addition, the FES system presented here has significantly lower power consumption and occupies a small volume. Hence, it is well suited for daily use in portable applications.

The key element of the presented FES architecture is a

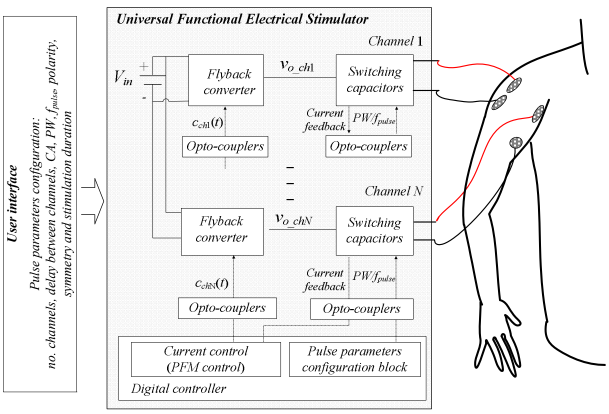

Fig. 2. Block diagram of the universal functional electrical stimulator.

novel digitally controlled multi-output power stage. It combines a single flyback and multiple switch-capacitor (SC) converters.

The flyback is used to step up battery voltage and for galvanic isolation. The downstream SC stages produce current pulses with a high current slew rate and, at the same time, ensure zero net charge of the stimulated cells. The operation of both stages is regulated by a digital controller, providing high flexibility and, hence, variety of the pulses. The controller also provides safety features, through multiple levels of over-current and over-voltage protection. Reference pulse patterns are provided through digital input setting the values for the desired amplitude, frequency, and durations of the pulses.

II. System Operation (Power Stage)

The operation of the universal FES system can be described by looking at Figs. 2 and 3, showing a block diagram of the system and its power stage, respectively. The user interface sets the stimulation parameters, such as, frequency, amplitude, pulse pattern, and the pulse duration. These parameters are fed to the digital controller, as reference signals for multiple parallel output power stages, as shown in Fig.3, which can be independently programmed. Each power stage consists of a flyback converter connected to the battery and a downstream switch-capacitor (SC) converter. The flyback steps up the voltage to up to 300 V, needed for a number of applications [1]-[13], and, at the same time, provides galvanic isolation required by safety standards [21]. The following SC stage provides high slew rate pulses and ensures zero net-charge to the cells. Those pulses are sent to the tissue through a set of surface probes.

The amplitude of the current pulses is regulated by both stages utilizing a new voltage-programmed current mode technique described in the following subsection.

A. Creation of zero-net charge high-slew rate pulses

Creation of zero-net charge current pulses with a high current slew rate has proven to be a challenging task [18], [20], [22]. As mentioned earlier, due to better controllability of the pulse amplitude current control of the pulses is predominantly used in FES systems. However, a drawback of this approach is slower slew rate in comparison with voltage mode solutions, causing more painful therapy. This is mostly due to the properties of the stimulated tissue, which can be modelled as a parallel combination of a resistor and a capacitor [23]. Also a related problem is accumulation of charge due to an imperfect shape of the bipolar pulses causing a difference between areas A_1 and A_2 of Fig.1. To eliminate the accumulated charge several solutions have been proposed in the past. Those include use of a resistive discharge circuit [24] that limits the maximum frequency of the pulses due to the extra time needed for the energy recharge. Also, a method [25] based on the use of a PI regulator has been proposed. While, theoretically, this method gives exceptional results in practice it can be affected by offsets of the Op. Amp. circuits and other system variations. Placement of a capacitor in series with

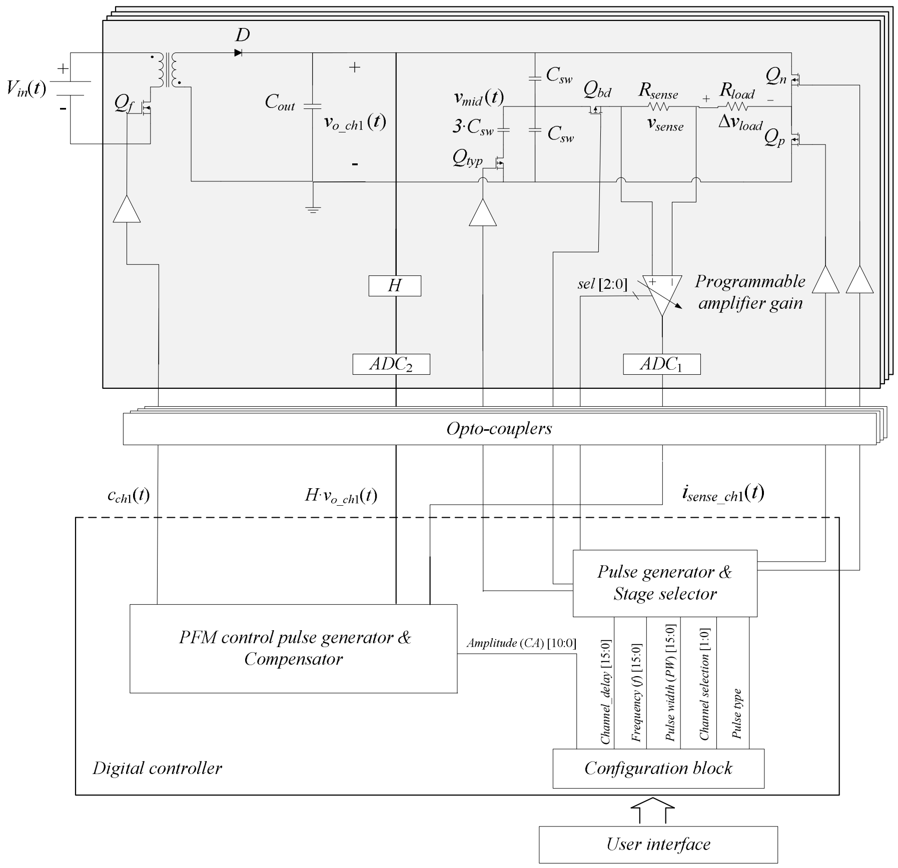

Fig. 3. Multi-output power stage of the universal functional electrical stimulator.

a current mode output stage [22] completely eliminates the net charge, but further reduces current slew rate, since the capacitive load is effectively increased.

To eliminate both of the previously mentioned problems a novel solution based on a SC circuit that is regulated by voltage-programmed mode current control is introduced here. As described in the following section, this control method is a dual of a well-known current programmed mode controller [26], where, in this case, the output current is controlled indirectly, by regulating the voltage of the stimulation probes.

The SC stage consists of a capacitive divider and a set of switches. The ratio of the divider can be changed between ½ and ¼, by turning on or off Q_{sym}. To describe operation of the SC stage we can use equivalent circuits of Fig.4, showing how an asymmetrical biphasic pulse (Fig.1) is created. During no output signal phase, all switches are turned off. To create the negative portion of the signal Q_n, Q_{bd}, and Q_{sym} are turned on. As a result, the output voltage of the amplitude $-4/5{\cdot}v_o$ is produced, where v_o is the midpoint voltage at the output of the flyback stage. The positive portion of the signal is created by turning on transistors Q_n, Q_{bd} and Q_{sym} and turning off Q_p. As a result a voltage of $+1/5{\cdot}v_o$ is created at the output. This phase is 4 times shorter than the negative phase. To create symmetric pulses, the divider ratio is set to ½. It should be noted that this operation has two advantages. Namely, *a voltage change with theoretically infinite slew rate is created and the tissue is always supplied through a serially connected capacitor, which blocks the dc components of the signal.* As a result, pulses with ultra-fast slew rate and net-zero charge are created.

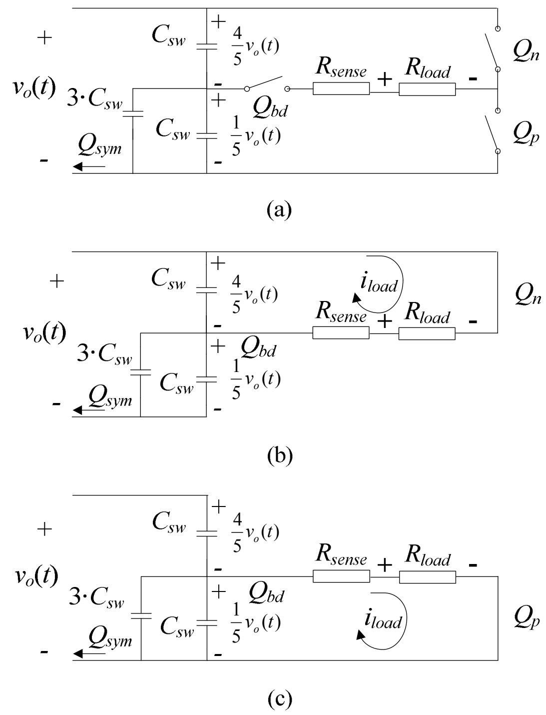

Fig. 4: Equivalent circuits of the SC stage during creation of a biphasic non-symmetric pulse. (a) No output. (b) Negative current phase. (c) Positive current phase.

B. Voltage-programmed mode current control

The amplitude of the pulses is regulated through voltage-programmed current mode control. This control method is a dual of the well-known current-programmed mode [26]. In this case, the amplitude of the output current is controlled indirectly, by changing the voltage at the input of the SC stage, i.e. by regulating the output of the flyback of Fig.3. The load, i.e. stimulating current is measured, by a sensing resistor and a programmable amplifier. This analog value is then transferred into a digital equivalent, by the ADC_1. This resulting output is compared to the digital current reference, i.e. desired stimulation current, supplied through the user interface. Based on the difference between two signals the output of the flyback v_o is changed. This change is performed through the pulse frequency modulation control scheme [27], to reduce switching losses of the converter. The ADC_2, measuring the mid-point voltage, is used for monitoring the safe operating range only.

As it will be explained in the full version of the paper, the role of the programmable gain amplifier is to minimize relative error in the current measurement caused by to quantization effects of the ADC_1. For low output currents the amplifier gain is increased, so the full range of the ADC is utilized and the quantization effects minimized. The full version of the paper will also give more details on the controller implementation, including realization of multiple overvoltage and overcurrent protection features and selection of the capacitors for SC stage based on maximum current limitations.

III. Experimental Results

Based on the diagrams of Figs. 2 and 3, a four-channel experimental prototype is created and tested with able bodied individuals. The specifications of the prototype are given in Table I. The specifications are set such that a broad range of applications is covered, from nerve (i.e. nerve cuff type or intramuscular stimulation) to denervated muscle stimulation. The requirements are obtained from FES practitioners and researchers that presently utilize multiple FES systems.

TABLE I
System Specifications

	Specification
Variable current level	0.5 mA -125 mA . Resolution = 0.5 mA
Variable pulse frequency	1 Hz - 15 kHz
Variable pulse duration	10µs - 10ms. Resolution = 1µs
Fast pulse rise time	10 ns
In balanced pulses, zero net charge	No residual charge left in tissue after each pulse
Variable pulse profiles	Biphasic Symmetric
	Biphasic Asymmetric
	Monophasic positive
	Monophasic negative
Voltage and current protection	Must be able to shutdown if voltage or current exceeds allowable limit
Multiple stimulation channels	4 stimulation channels
	Regulation of multiple channels independently

To verify operation of the system, comparative tests were conducted, where the operation of this system is verified on a test subject and also compared with a state of the art system [12]. The test with able bodied individuals was done with 5 human subjects between 20 to 40 years of age.

Test results of the new stimulator are shown in Figs. 5 to 7. Fig. 5 verifies independent control of four channels and clean noise-free, zero-charge pulse shapes. The tissue in the trial was exposed to stimulation pulses over long periods and no charge accumulation has been noticed.

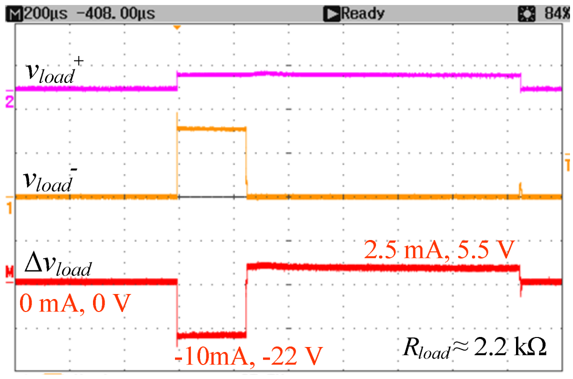

Fig. 5. Voltage pulse waveform, PW= 250µs, f_{pulse}= 45 Hz. Ch1: negative terminal of load voltage, 20V/div. Ch2: positive terminal of load voltage, 20V/div. M: Ch2-Ch1.

As it can be seen from Fig.6, showing zoomed in transition, rise time is approx. 10 ns. In comparison with the [12], it is about 150 times faster. As a result, the subjects reported experiencing less pain sensation during the sessions.

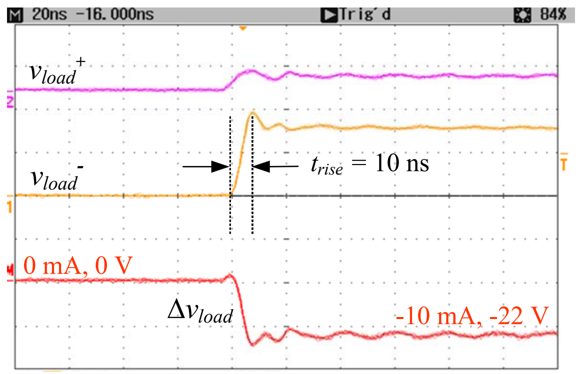

Fig. 6. Rise time waveform. Ch1: negative terminal of load voltage, 20V/div. Ch2: positive terminal of load voltage, 20V/div. M: Ch2-Ch1.

Figure 7 shows the torque exertion in biceps femoris stimulated by the new stimulator and by [12]. The torque was measured using the Biodex-System 4 Pro [28]. It can be seen that, for the same torque, the new stimulator requires about 34% lower current (18 mA compared to 27 mA). Consequently, the life span of the battery is increased. In the full version of the paper, additional experimental results will be provided as well.

IV. Conclusions

A novel architecture of a highly-flexible universal current-mode functional electrical stimulator is introduced. The key elements of the stimulator are novel output power stage and a complementary controller. The power stage combines a flyback and switch-capacitor

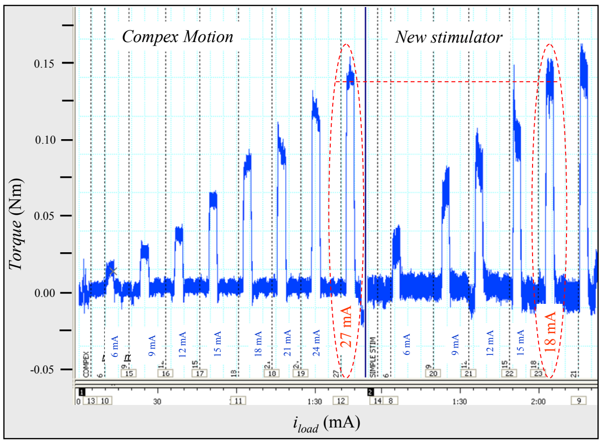

Fig. 7. Torque exertion (biceps femoris).

(SC) converters, which is regulated by a digital voltage-programmed current mode control. The combination of the control method and SC output provides wide variety of pulses, ultra-fast current slew rate, and inherent zero charge build up. These result in a less painful therapy and longer battery life allowing the functional electrical stimulation system to be used as a portable device on daily basis. Comparative clinical trials with a state of the art solutions verify drastically improved current slew rate and about 1.5 times lower pulse amplitude needed for the same muscle contraction. Finally, the battery life is significantly increased due to the novel functional electrical stimulator requires 60% lower energy.

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