DEVELOPMENT OF AN INTEGRATED ELECTRICAL STIMULATION SYSTEM WITH FEEDBACK FOR PHYSICAL REHABILITATION

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Abstract: In physical rehabilitation, electrical stimulation is widely used as a therapeutic method. However, as it is not common to find portable devices, capable of integrating information from different sensors, and also with flexibility in signal generation and triggering. This paper presents an integrated electrostimulation system that encompasses all those facilities. The system integrates feedback signals coming from an accelerometer and is capable of adapting electrostimulation depending on motor performance. The device uses a microcontroller for the waveform generation, and allows controlled waveforms to be produced in response to signals read from feedback sensors. Besides this high versatility, the principle of the power generation employed by the device and additional hardware circuitry also provides mechanisms to ensure patient safety in the unlikely cases of malfunction of the microcontroller. Here we also present an example of application of the device that uses real time feedback information to control electrical stimulation.

1 INTRODUCTION

The latest technological development in electronic miniaturization opens new perspectives in the development of more sophisticated systems integrating multiple components. Electrical stimulation (ES) devices for medical rehabilitation could benefit from these advances. Computational power has also been made widely accessible, with the appearance of highly portable inexpensive computers. ES devices could use evolved computational models to dynamically control the electrical stimulation delivered by several independent channels, responding to multiple sensors, all working synchronously and according to patient specific rehabilitation programmes.

Medical rehabilitation uses electric stimulation intensively (Hennings et al., 2006). Usually, patients with mobility limitations suffer from muscle atrophy, which hinders their recovery. Electric stimulation is an artificial way of inducing motor movement and to prevent atrophy (Buckley et al., 1987; Langzam et al., 2006; Durfee, 1999). It can lead to synergistic gains, as the recovery of some muscle fibbers allows the recovery of many others. Indeed, clinicians believe that considerable functional recoveries can be induced through exercise (Doucet and Griffin, 2008). One main problem, however, has always been in how to induce physiological movements. This can be extremely relevant as otherwise inadequate reinervation may be stimulated leading to poor motor functional recovery (Al-Majed et al., 2000; Franz et al., 2008). Inducing physiological movements through ES has been difficult for two main reasons. The first is that electrical stimulation tends to produce non-physiological muscle activation. Secondly, most equipments do not use feedback information to produce stimulation.

The minimisation of these difficulties motivates the present work. Here we discuss the development of a small, light and low power consumption circuit that provides a closed loop electrical stimulation. The circuit is capable of acquiring data from sensors (like accelerometers, force sensors and alike), communicating with a computer which can respond through a versatile real time activation of ES stimuli. User performances are registered and can be used in later clinical evaluation.

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This article will focus on the development of an ES circuit capable of producing symmetric biphasic pulsed current stimuli, with real time control over ES parameters (pulse phase, frequency and current modulation). The circuit uses a microcontroller (PIC) to integrate ES with inertial sensors data acquisition and communication with a computer. Before discussing the circuit, it is important to introduce the current terminology used to characterize electrical pulses in ES devices, and also to discuss what their impact is in ES therapies. This is done in the next section.

2 ELECTROTHERAPEUTICAL REQUIREMENTS

Electrical stimulation is an artificial way to trigger Action Potentials through the application of rapidly changing electric fields on excitable tissues. Transcutaneous Electrical Stimulators are the most common (Nelson et al., 1999). They apply currents through the skin to excite peripheral nerves.

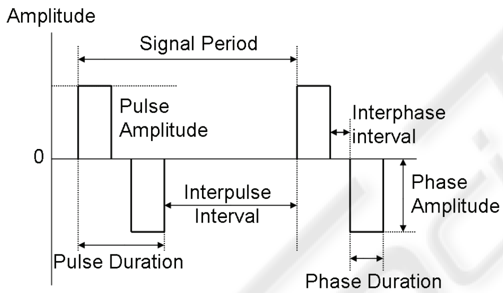

Figure 1: Parameters defining a symmetrical biphasic pulse.

Pulsed current waveforms are classified in monophasic or biphasic. In monophasic pulse signals current flows only in one direction (one phase) between the electrodes. In biphasic signals each pulse has two opposing phases. Biphasic pulses can be symmetric or asymmetric, being symmetric variants more appropriate for clinical applications since they leave no residual charges in the stimulated tissues.

Biphasic pulses are characterised by several parameters (Figure 1): the pulse amplitude in each phase, the interphase interval separating each phase, the interpulse interval and the pulse duration.

Pulsed current signals can be modulated, varying pulse duration, peak amplitude or pulse rate. This modulation can be in the form of bursts, interruptions and ramping. Bursts are created by trains of pulses that flow for some milliseconds and then stop for another period of time in a periodic way. Interrupted pulses are a sequence of pulses interrupted for a period of time for resting. Interrupted pulses promote only twitch (brief) contractions. For higher ES frequencies, bursts fuse leading to tetanic contractions. In this case the period between pulses is shorter than the muscle contraction-relaxation cycle, summating the forces produced by each impulse (Kitchen, 2002).

The muscle contraction strength can be augmented by engaging more motor units or by increasing the frequency at which each motor unit fires (Vrbová et al, 2008). A higher number of activated motor units can be achieved by applying pulses with higher amplitude. The force produced by increasing the ES frequency depends on the fibre type. There are two main fibre types in a muscle, type I and type II. Their proportion varies from muscle to muscle and depends on function. Fibres can change their type with exercise, and this is particularly important in the medical rehabilitation context. Indeed, in patients without active voluntary movement, typically they loose type I fibres, a process that can be reversed and/or prevented by ES.

Type I fibres are slow twitch muscle fibres that produce relatively low forces but are also the less fatigable. They are particularly important in maintaining posture. As a result of their slow contraction and relaxation they fuse at lower frequencies when compared with type II fibres. The later, are responsible for fast movements, contracting and relaxing faster and consequently fusing at higher frequencies. It is typically the aim of the therapist to select the ES programme that best selects the fibre type to be enhanced.

3 DEVELOPMENT OF AN INTEGRATED ES DEVICE

The literature on ES circuits is not abundant although a few examples can be found (Cheng et al., 2004; McPartland and Mook, 1995). To deliver an ES stimulus, any ES device has a voltage elevation stage required to overcome the high skin and other tissues impedance.

Some methods to increase voltage use transformers, charge pumps, or switch mode power supplies. Transformers are costly and large and on high frequencies, inductive reactance grows decreasing their efficiency. Charge pumping is another way to step-up voltage with capacitors and switching elements. There are several charge pump configurations, like the Cockcroft-Walton voltage multipliers and the Dickson charge pump (Pan et al., 2006). The strategy works by charging capacitors in parallel and then, using switches to rearrange the circuit, discharge the capacitors now in a serial configuration.

Switch mode power supply (SMPS) is a power conversion where high frequency switching is used. Intrinsic spurious capacitors/inductors in some electrical components may be used as more effective means to reduce size and take advantage of their effects. A particular interesting topology is the quasi-resonant converter, which has the advantage of reducing switching losses (Ye et al., 2004; Pressman, 1998). This is performed by adding a resonance inductor in series with a switch, a catch diode and a resonance capacitor, as illustrated in Figure 2. When the switch is on, the inductor stores energy in its magnetic field. When the switch is turned off the stored energy resonates with the capacitor. The capacitor is charged with a sinusoidal waveform, as shown in the V_{SWITCH} curve in Figure 2 (right). The diode stops resonance after half period. This is valid for high switching rates (within the half resonance period) as the resonance does not have time to resume. In this case, there is zero voltage across the switch (ZVS). Switching loss is thus reduced as the voltage across the semiconductor device is set to zero before switching (Neacsu, 2006). This circuit requires an on-off time control that can be easily handled using a microcontroller.

Microcontrollers have many advantages, such as the availability of analogue to digital converters and acquisition modules that allow following in real time the stimulus amplitude delivered to the patient and acquire feedback information from sensors. Also, communication modules allow information display, remote control of peripheral components such as digital potentiometers, read information from digital sensors and interact with personal computers or other processor units. Timer modules help designing the waveform signal for the ES program.

The microcontroller improves the ES functionality as it allows programming and reprogramming, to change parameters according to a pre-established algorithm, to give instructions to the therapist and, of uttermost relevance, they can be used to control safety levels. These were strong motivations to use a Peripheral Interface Controller (PIC), commonly abbreviated as “microcontroller” (microchip PIC model 18f25K20), in the development of the presented electrical stimulator system.

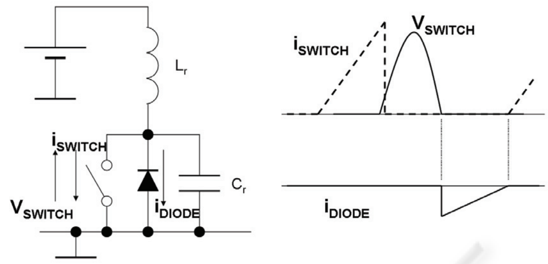

Figure 2: Zero Voltage Switch circuit and waveforms.

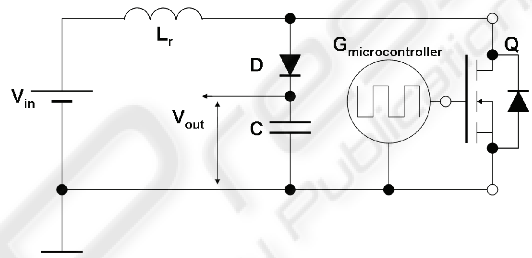

Figure 3: Circuit diagram of the switching resonant voltage elevator.

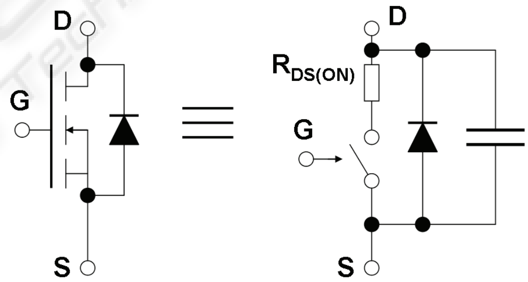

Figure 4: Simplified equivalent circuit for a MOSFET.

The proposed ES circuit uses a switching resonant voltage elevator (SRVE) to generate voltage pulses, as shown in Figure 3. The resonance tank is composed by an inductor L_R and the MOSFET parasitic capacitance shown in more detail in Figure 4. Hence, the resonating capacitor is already built-in in the MOSFET. The circuit in Figure 3 applies the ZVS method, in a similar way to the circuit displayed in Figure 2. The MOSFET performs the switching, but it also includes the diode and the capacitor, so that the circuit indeed performs the ZVS as the one in Figure 2.

The voltages across the capacitor and inductor depend on the current supplied to the inductor during the switch-on period, and also on the switching frequency and on the inductance and capacitance values. Large voltages can thus be generated with low power consumption and a relatively small voltage power supply.

The high voltage generated during the switching is then stored in the large capacitor, C, in Figure 3, which can then be applied to the patient. Naturally, the charge stored at each switching moment is limited by the supplied power. The voltage is sustained through the continuous generation of high frequency voltage spikes controlled by the microcontroller.

Another advantage of using high frequency switching is that it minimizes components size and cost. One example is the advantage taken from MOSFET parasitic capacitances.

On the other side, it could be argued that high frequency signals (the best results were obtained with 1MHz switching frequencies) could suffer from electromagnetic interferences. However, resonant circuits produce sinusoidal currents which have smooth changes, and hence are less prone to this type of problems.

It should be noted that there are already some circuits in the literature that use similar strategies for voltage elevation (Cheng et al., 2004). However, the solutions presented do not consider biphasic pulses and furthermore, require using two independent switching elements. This makes the circuit more complex, possibly with no functional advantages.

In the present circuit, output voltage is not directly applied to the patient. Instead, it is directed onto a subsequent circuit (the waveform driver) as depicted in Figure 5. This circuit is based on an H-bridge, and was designed to control the duration and polarity of the current that passes through the load resistance, R_{LOAD}. Control is made by two microcontroller output signals, G_1 and G_2. At all moments, only one of the G_1 or G_2 signals can be in a high state. With this procedure, biphasic pulses can be generated since the current flows on different senses through R_{LOAD}, depending on which signal G_1 or G_2 is active. It is also possible to generate symmetric or asymmetric pulses by changing the duration of each G signal in the high state. The PIC timer modules control pulse duration, pulse frequency, and activation/rest periods of the stimulus application.

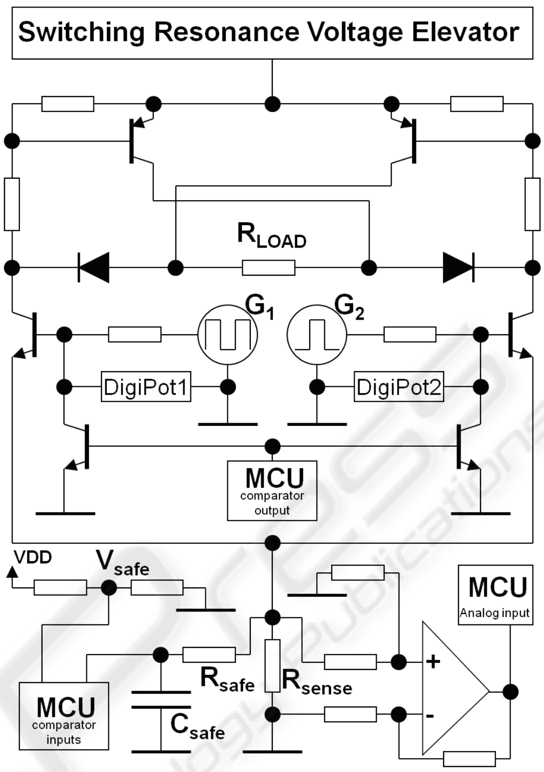

Figure 5: Scheme of the ES circuitry. It includes the SRVE, the waveform driver, the current sensing unit and the injected charge safety control circuits.

To stimulate excitable fibbers in a selective way and to prevent damaging tissues in the stimulated area, the charge flow must be controlled. As living tissues impedance is variable, the most effective and safe strategy, is to control waveforms in current rather than in voltage (Merrill et al., 2005).

The circuit that controls phase amplitude operates in a closed loop. The signals from G_1 and G_2 are voltage divided by a resistor and a digital potentiometer, as represented in Figure 5 by DigiPot1 and DigiPot2, respectively. These digital potentiometers are adjusted in real-time by the PIC microcontroller, controlling the voltage at the MOSFET gate and limiting the maximal current through R_{LOAD}. This current is measured by an analogue input to the PIC, after amplifying the voltage drop across the small R_{sense} resistor.

The presented ES system incorporates a circuit to limit the charge injected for safety purposes. The charge increases with the amplitude of the current and the phase duration. The maximal current allowed was designed taking into account the electrode-tissue interface area to avoid hazardous charge densities (Shannon, 1992). This depends on the size of the electrodes used. The capacitor C_{safe} in Figure 5 is charged by the small voltage drop across R_{sense}. The comparator module of the microcontroller used is programmed to automatically shutdown the PWM signal used in the SRVE when the voltage across C_{safe} reaches V_{safe}. At the same time, the comparator output pin pulls down the gates of the waveform driver circuit to values below the conduction threshold. In case of undesired charge densities, the voltage elevation and the current flow across R_{load} stops immediately.

Feedback information is measured by a 3-axis digital accelerometer MMA7456L manufactured by Freescale Semiconductor. This sensor is applied in functional electrical stimulation.

4 EXAMPLE OF APPLICATION

In this section we present an example of application of the feedback ES device described above. The prototype used is shown in Figure 6. The circuit is powered by a 9V battery.

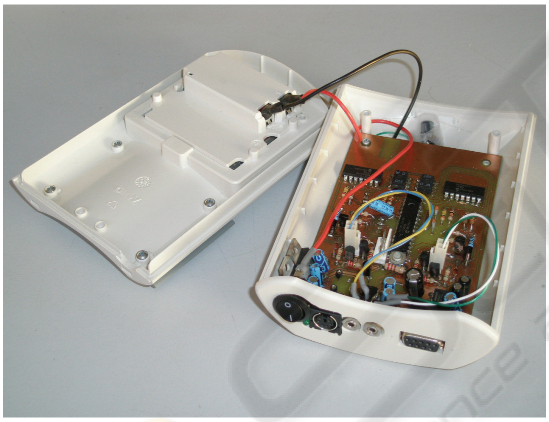

Figure 6: Image of the electrostimulator prototype.

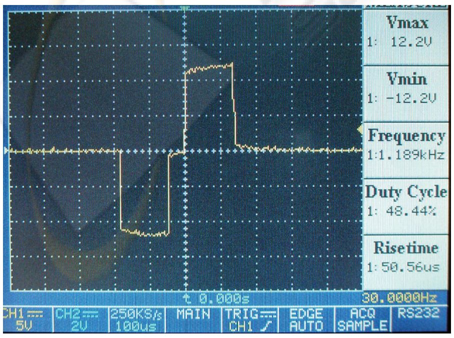

Figure 7: Snapshot of the oscilloscope monitor displaying a biphasic pulse as obtained experimentally with the prototype. 10x amplified probes were used in this picture.

Figure 7 shows a biphasic pulse obtained experimentally with this prototype. For the purpose of this figure a load resistance of 10kΩ was used. The pulse is symmetric and biphasic. It has a pulse duration of 300µs, an interphase interval of 50µs and 12mA of phase amplitude.

For a matter of illustration of the prototype at work we designed an experimental protocol that consisted of applying ES stimuli on the biceps of healthy subjects to produce non-voluntary movements. It was defined that the amplitude of the stimulation should never exceed 17mA. Furthermore, the amplitude of the ES should decrease anytime the arm reached a maximal angle of 83º (see Figure 8) and it should increase again if the arm fell below 41.5º.

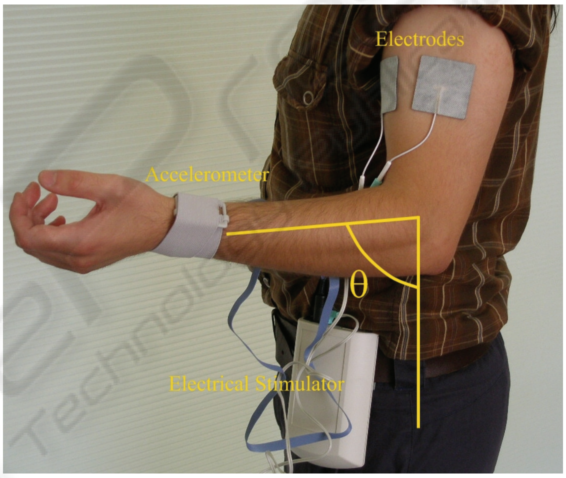

Figure 8: The ES exercise applied to the biceps of a healthy subject. In this figure we also show how the amplitude of the arm movement was measured has a result of the applied electrical stimulation.

Typical examples of the results obtained are shown in Figure 9. In the two exercises reported different ES frequencies were applied. It can be observed that the amplitude of the electrical stimulation never exceeds the pre-defined threshold value and that the evolution of the electrical stimulation changed whenever the arm overcame the maximal predefined amplitude of the movement. These results clearly show a frequency dependent muscle response that can be characteristic of muscle fibre type composition which can be valuable for clinical purposes.

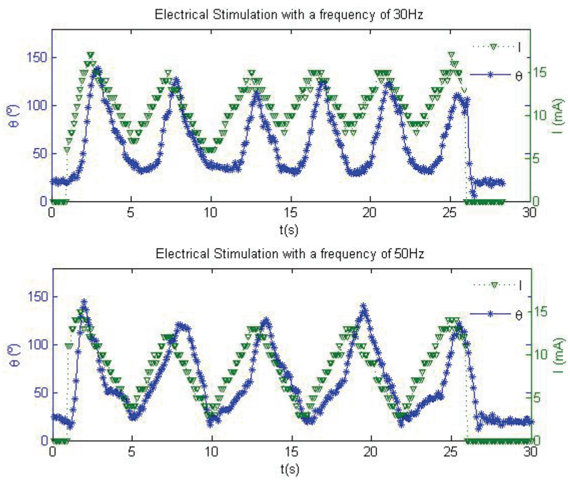

Figure 9: The amplitude of the arm movement, θ (in stars), and of the applied electrical stimulation (triangles) as a function of time. The stimulation frequencies were 30Hz *(Top)* and 50Hz *(Bottom)*.

5 CONCLUSIONS

This article presents an inexpensive small-sized transcutaneous electrical stimulator unit capable of integrating feedback information arising from a digital accelerometer with electrostimulation. The delivered pulsed currents can be controlled in real time and according to a programmed protocol, on a number of parameters - pulse duration and amplitude, pulse rate, the type of current modulation (burst, interrupt or ramp modulation).

In the future more elaborate ES programs should be developed to deliver optimal ES adjusted to a patient's needs. These programmes can use higher computational power and sophisticated theoretical models of muscle function, to design the most efficient ES programmes.

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