DESIGN AND TESTING OF AN INSTRUMENTATION SYSTEM TO DETERMINE THE EFFICACY OF PAIN REDUCTION DURING MUSCLE STIMULATION

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ABSTRACT

Functional Electrical Stimulation (FES) has been used for decades to restore muscle function following neural trauma. Another promising use has been to maintain or increase muscle strength following injury. Unfortunately in the latter case there is considerable stimulation pain for the sensory intact subject during effective levels of stimulation using surface electrodes. Recent research [1,2] has suggested using a constant long (up to 10 msec) low amplitude or ramped conditioning pulse just prior to the high amplitude stimulus pulse (100 – 200 μsec) will reduce the excitably of sensory nerve fibers. However, commercial muscle stimulators cannot be easily modified to provide such complex pulse patterns and flexible pulse train control. We have designed and implemented a novel very flexible LabVIEW based monophasic constant current muscle stimulator that provides pulse trains with long duration pre-pulses and high voltage stimulus pulses with selectable shapes, amplitudes, durations and frequencies. The stimulator hardware uses a standard voltage to current converter circuit with an efficient high voltage DC/DC converter which can present up to 100 ma pulses to surface electrodes. As well, the stimulator system includes an isolated EMG amplifier to record the evoked M-waves, which are used to estimate the fraction of muscle motor units being stimulated. The system is presently being tested in a study to determine the efficacy of the pain reduction stimulation strategy described above.

INTRODUCTION

Electrical muscle stimulation has been used for decades to improve muscle strength or provide function in the rehabilitation population. However, effective levels of stimulation result in excessive pain in the sensorially intact subject. Promising research has been done in using conditioning prepulses to selectively recruit smaller motor units in rat skeletal muscle [1]. As well, these prepulses have also been used to raise the pain threshold when stimulating peripheral sensory fibers in humans [2]. The overall objective of our research is to investigate whether conditioning prepulses can be used to decrease the level of stimulus pain during high levels of repetitive muscle stimulation to improve the muscle strength of patients with chronic obstructive pulmonary disease (COPD). This paper presents the design and testing of a new computer controlled stimulator that can deliver complex stimulus waveforms. This instrument will then be used to test the hypothesis that pre-stimulus conditioning pulses are effective in reducing stimulus pain, and which waveforms are best to achieve high levels of stimulation coupled with maximum pain reduction. Since commercial stimulators do not provide adequate pulse width, duration and shape control, a custom design was pursued. Previous studies utilized high voltage designs but were strictly limited to low current and very high impedance loads [3]. Our design encompasses currents up to 100mA with load impedances in the range of 2K ohms.

SYSTEM DESIGN

Description

The block diagram of our system is illustrated in figure 1. Each block is a separate module and can be used independently provided ±5V is supplied to the module. This allows for rapid prototyping and in the case of failure or future upgrades, minimal effort is needed to obtain full operational status again.

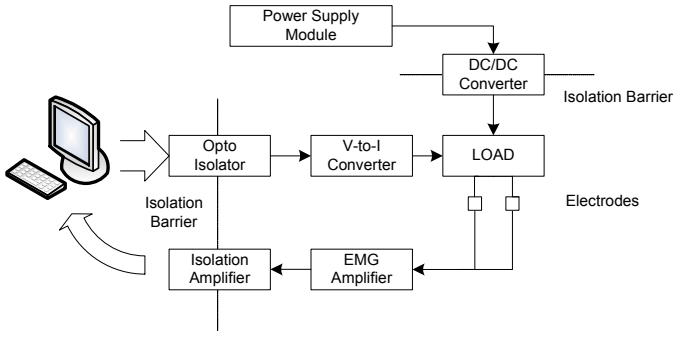

Figure 1: Block diagram of overall system design

All stimulus parameters are controlled using a PC running the LabVIEW software package. The software controls the frequency, duration, amplitude, ramp up and ramp down times of, the stimulus pulse, prepulse and the pulse trains.

Isolation between the patient and the system is provided via a number of isolation barriers. The high voltage DC/DC converter provides an isolation rating of 1500VDC. Control signals from the PC are isolated using an analog optoisolator (HCNR 201). Input signals from the EMG module are isolated using a standard isolation amplifier (AD 202) set as a unity gain buffer. Both the EMG module and the current control amplifier are powered using separate miniature DC/DC converters from Texas Instruments (DCH010505D) which allow for isolation of the system ground plane from the AC ground plane.

Output Stage

A simple voltage-to-current conversion circuit is utilized to provide a constant monopolar output current. It is based around a high voltage transistor driven by an op-amp as shown in figure 2. An isolated control signal from LabVIEW sets the stimulation amplitude and duration. A gating transistor, Q_1, on the high voltage end protects the patient from stray currents when stimulation parameters are set to zero. It can also be used to implement an H bridge configuration for bipolar stimulation.

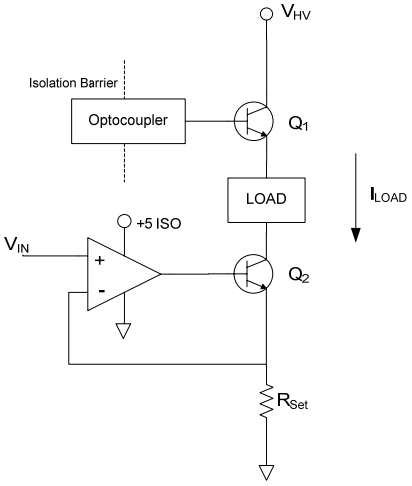

Figure 2: Constant current source

The operation of this circuit is straightforward. Assuming base currents are negligible, a voltage V_{IN} is impressed on the positive terminal of the op-amp which forces a current across R_{SET} that is equal to V_{IN}/R_{SET}. As this is the emitter current (I_E), the collector current of Q2, which is that of the load, will be very similar ($I_C \sim I_E$). Therefore, the output current is directly proportional to the input voltage V_{IN}.

$$I_{LOAD} = I_C \approx V_{IN}/R_{SET} \quad (1)$$

However, when base current flowing through Q_2 is considered, an output current error is present and is shown in (2).

$$I_{ERROR} = I_C/ß \quad (2)$$

This error is proportional to the ß value of the transistor which in our case is approximately 30 (TIP 50).

The limiting factor of this voltage-to-current converter is its voltage compliance. Currently, we are using a 200V, 4W, DC/DC converter (PICO12VV200S), which for our purposes provides adequate voltage to drive our impedances at the desired current levels.

System Software

Custom software was written to control various settings as shown in figure 3. The software allows for control of prepulse amplitude, duration and type; stimulus pulse amplitude and duration; ramping up and down times as well as plateau and off times. On the fly changes to these parameters can be made easily and rapidly through this interface.

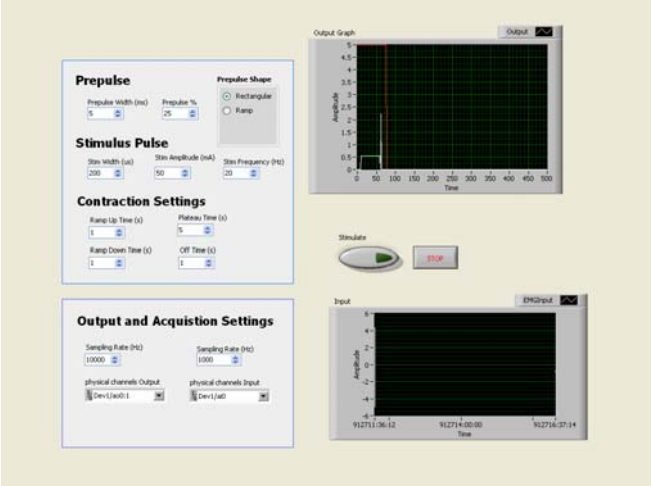

Figure 3: Front panel view of LabVIEW software interface

HARDWARE TEST RESULTS

To test the stimulator a resistive and capacitive load was placed across the output of the stimulator to simulate electrode and skin impedances. Both ramped

and square prepulses of various durations and amplitudes were tested and waveforms were recorded using an Agilent 54621A Oscilloscope. Typical results from these tests appear in Figure 4. At present, only ramped and rectangular prepulses are programmed into the software but any waveform shape is attainable.

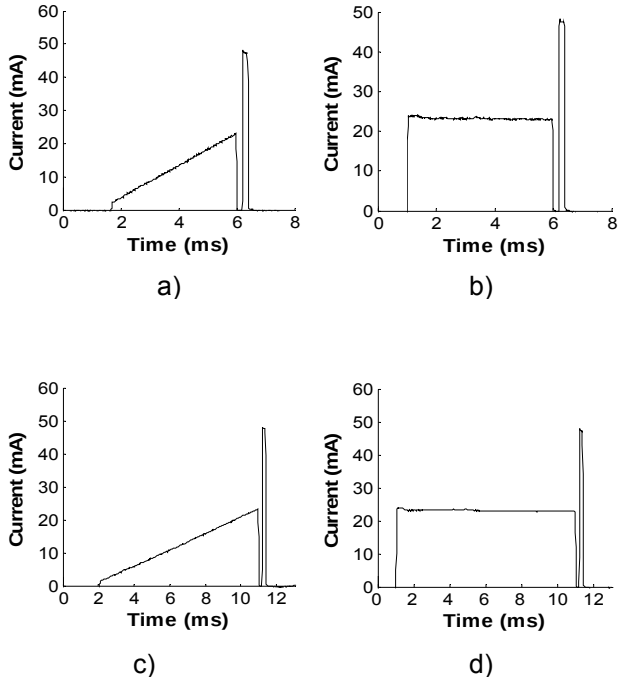

Figure 4: Stimulus current waveforms for 5ms ramped and rectangular prepulses, a) and b), and for 10ms ramped and square prepulses, c) and d).

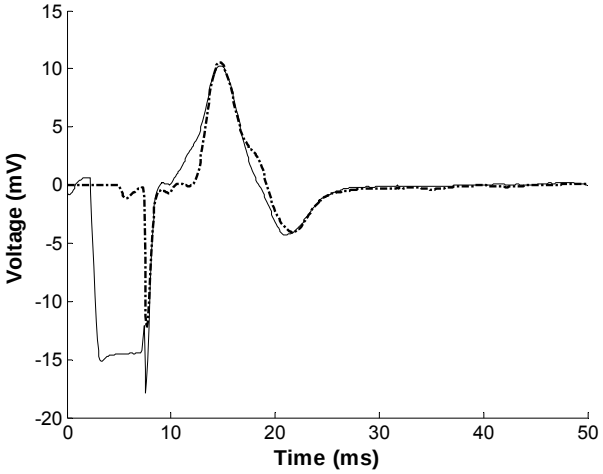

Figure 5: Typical M-waves recorded from the thenar muscle during stimulation with and without a prepulse

ELECTROPHYSIOLOGICAL TEST

To test the isolation of the various modules and the operation of the EMG amplifier, it was necessary to instrument a subject and record the evoked potential (M-wave) from a selected muscle. It was decided to stimulate the median nerve at the wrist with 100 μsec pulses using large (4 cm X 4 cm) graphite impregnated rubber electrodes, and record the M-wave from the thenar muscle. These electrodes are the ones that will be used in our COPD muscle strengthening research studies. The recording electrodes were 27mm by 11mm (Sentry Medical Products, Irvine, CA) with the stigmatic electrode placed over the thenar eminence to cross the first metacarpal bone perpendicularly at the junction of its proximal and middle thirds as shown in Figure 2. The reference electrode was attached to the proximal phalanx of the thumb. A ground electrode was located at the dorsum of the hand. The voltage gain was set to 250 and the signal was hardware bandlimited to 3 to 500 Hz.

Figure 5 shows typical near maximum M-waves recorded for stimulation without any prepulse (dashed line) and with a 5 ms prepulse at 20% of the stimulus pulse amplitude (solid line) for the same stimulus pulse amplitude. This figure shows that the stimulus voltage artifact for the 5 ms prepulse closely mirrors the original 5 ms current stimulating waveform while that for the 100 μsec stimulating pulse is attenuated because of the hardware filters.

DISCUSSION AND CONCLUSIONS

The new instrumentation system meets the design criteria, is easy to use, and safe because of the extensive isolation. We have only tested ramped or constant amplitude pre and stimulus pulses but the design allows one to select and implement any pulse shape. Figure 5 shows that even when the recording electrodes are in close proximity to the stimulating electrodes (as will be the case in our clinical research), the stimulus artifact is small and the full M-wave can be easily measured. Of special note in Figure 5 is the fact that the two M-waves have nearly identical shape indicating that the long duration prepulse did not stimulate any α motor neurons or inhibit any from being depolarized by the following stimulus pulse. This is important since we do not want inhibition of motor neurons during muscle strengthening but only of sensory fibers.

We will now be conducting several studies to determine the amplitude effects and relative merits of using ramped or constant amplitude prepulses as

large axon inhibitors [1] during surface muscle stimulation. As well we will be using the system to determine the efficacy of prepulse inhibition of sensory pain fibers.

ACKNOWLEDGEMENTS

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[2] C.J. Poletto, C.L. Van Doren, "Elevating pain thresholds in humans using depolarizing prepulses," *IEEE Trans. Biomed. Eng*, vol. 49, no. 10, 2002.

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