A Programmable Electronic Stimulator for FES Systems

Milan Ilić, Dragan Vasiljević, and Dejan B. Popović, *Member, IEEE*

***Abstract*— A low-cost, programmable, galvanically isolated bipolar four channel stimulator for patterned electrical stimulation has been developed. The design is based on a newly developed concept, and it includes a high efficiency DC/DC converter securing high output voltage and sufficient energy to drive four output stages, a microcontroller for control of charge compensation, pulse rate and duration, as well as patterns of exercise and functional use regimes. A user-friendly interface for programming of the stimulator has been developed for the PC host computer, using Windows environment and a push-button control for functional application of the system. The use in humans who suffered a central nervous system lesion and lost voluntary control over muscles necessary for standing, walking, or manipulation was tested. The prototype is small, light, battery-operated, and flexible for various applications.**

***Index Terms*—FES, stimulator, power source.**

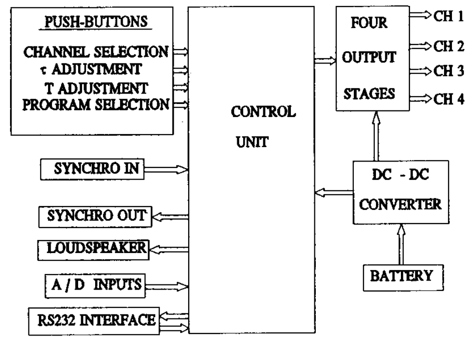

Fig. 1. The block diagram of the stimulator.

I. Introduction

NUMEROUS stimulators have been designed for surface stimulation of neuro-muscular structures, from simple single-channel, to very sophisticated multichannel units [1]–[7], [9], [10], [12], [14], [21]. Several commercially available stimulators ((2Ch)-Respond II, Medtronic, USA; (2Ch)-Jožef Štefan, Ljubljana; (2Ch)-Bioengineering Unit, Strathclyde University, Glasgow; (4Ch)-Parastep-1, Sigmedics, Chicago, IL, (4Ch)-DES, Novi Sad, Yugoslavia, (4Ch)-Quadstim, Biomech Design, Edmonton, Alberta), developed for standing and gait restoration, have been used in humans suffering paraplegia. It became obvious that each of these systems is meant for a specific application; hence, it was almost impossible to use them differently than suggested by the manufacturer.

Two goals were adopted for this project: development of a new concept for a programmable stimulator for FES and the design and testing of the prototype. The new concept involves the operation of the stimulator in two modes: the Setup Mode (SM) for programming and setting the parameters of stimulation and the Autonomous Mode (AM) when the stimulator generates the programmed stimulation pattern (Fig. 1). The following features were adopted: 1) four-channel, programmable, portable and battery-powered; 2) bipolar, compensated monophasic stimulation pulses, obtained via optically isolated, constant-current output stages; 3) programmable stimulation patterns within the following limits: pulse duration $\tau = 10$–500 μs, step 10 μs; pulse rate (frequency) $f = 1/T = 5$–99 Hz, step 2 Hz; maximum positive going pulse amplitude $I = 0$–140 mA, step 2 mA [11], [18]; 4) availability of a synchronization output and input pulses; 5) low power consumption; 6) low cost; and 7) a control unit suitable for nonexpert programmers and easy operation.

System requirements comply with the following findings: 1) A minimum of four channels of FES is required for ambulation for a subject with a complete motor lesion of the lower extremities and preserved balance and upper body motor control [8]. A device having up to four channels of stimulation can be used in subjects with an incomplete spinal cord lesion, expressing functional deficit in several muscle groups [19], [20] but still being able to stand and ambulate with great difficulty. 2) The use of several analog and digital transducer inputs for the control can improve gait by greatly reducing the duration of the stance phase and taking better advantage of gravity and inertial forces [15]. 3) Four programmable channels can be implemented for the control of triceps and biceps in the upperarm, forearm wrist extensors and flexors, and thumb extensor and flexor. An interesting example of such an application is for the control of reach in humans who suffered spinal cord injury at C5/C6 level [16]. 4) A flexible programmable stimulator is needed for a variety of training regimes and exercise routines.

This stimulator was designed for several possible applications: 1) restoration of walking in subjects with a complete spinal cord injury at the thoracic level (T5–T11); 2) restoration of walking in subjects with an incomplete CNS injury (C6–T11); 3) restoration of reaching and grasping for subjects

Manuscript received November 5, 1993; revised August 18, 1994.

M. Ilić and D. Vasiljević are with the Faculty of Electrical Engineering, Belgrade, Yugoslavia.

D. Popović is with the Faculty of Electrical Engineering, Belgrade, Yugoslavia, and The Miami Project, University of Miami, Miami, FL 33136 USA.

IEEE Log Number 9406448.

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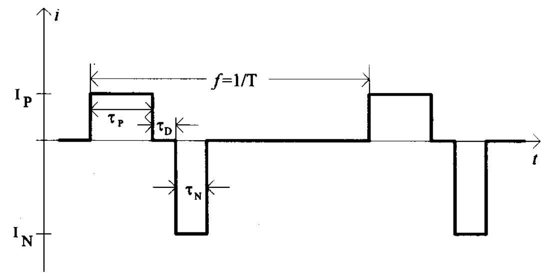


Fig. 2. The time diagram of output pulses. The pulse duration can be adjusted between 10 and 500 μs and the interpulse interval T between 10.001 and 200 ms. Note that the area under the positive pulse is equal to the area determined with the negative pulse.

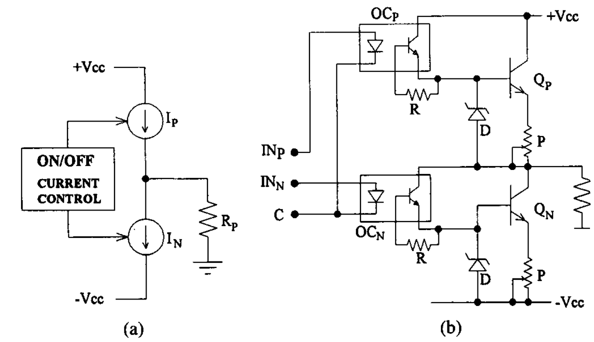


Fig. 3. The block diagram (a) and the schematic of the output stage (b). D-BZY4V1, Q_P, Q_N-MJE340, P = 500 Ω, OC_P, OC_N-H11D1.

who suffered a loss of arm and hand functions (C4–C6) using surface electrodes; and 4) exercise (programs for muscle strengthening, joint range articulation, etc.).

II. The Stimulator Design

A. The Output Stage

The output stage must: 1) generate constant current pulses of both polarities; 2) allow manual adjustments of the amplitudes of positive (I_P) and negative pulses (I_N) via potentiometers; and 3) control the duration of positive (τ_P), negative (τ_N), delay time between them (τ_D), and the frequency, $f = 1/T$, (Fig. 2).

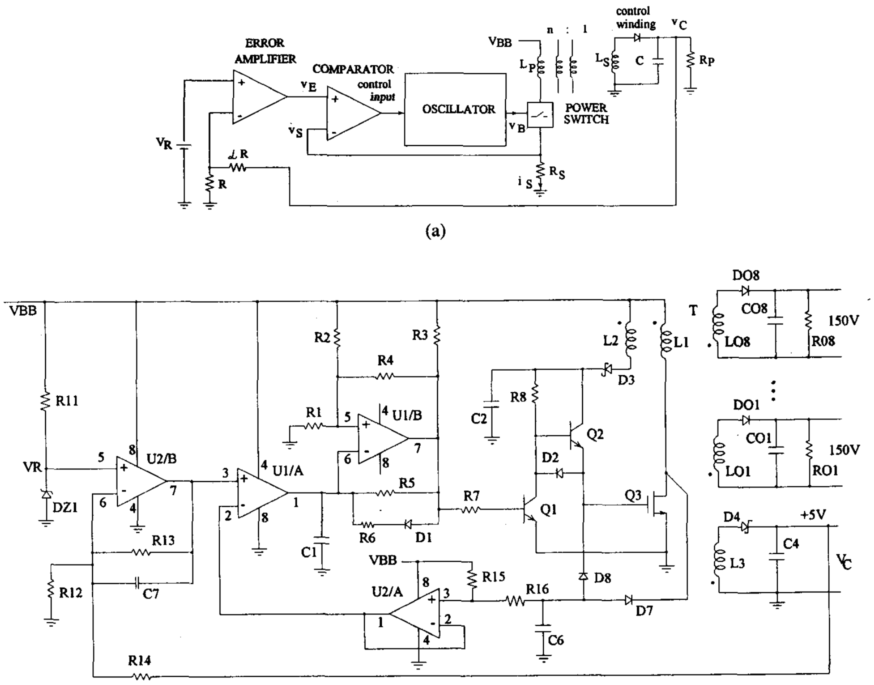
The output stage is realized with two push-pull connected constant current sources able to sink or source current to a muscle having electrical resistance R_P (Fig. 3(a)). Current sources are switched by galvanically isolated digital signals, which provide control of all output stages from the same control unit (Fig. 3(b)). The constant current source is designed with high voltage transistors, Q_P and Q_N driven by optocouplers, OC_P and OC_N. The amplitude of the constant current is determined by the Zener diode breakdown voltage and potentiometer resistance, P. The amplitude of the negative going current pulse is equal to the collector current of transistor Q_N (Fig. 3(b)). The positive going pulse contains two components: the constant collector current of the transistor and the current through the Zener diode determined by the optocoupler photo transistor. The second component of the current is not constant due to the Early effect of the optocoupler photo transistor. Therefore, muscles with various resistances, R_P receive current pulses having different amplitudes varying within $\pm 10\%$ about the expected current value. The described variability is acceptable for this application. The proposed output stage is physically small, simple to build, and reliable. The standby power consumption is zero when the load current is zero. The temperature stability of the pulse amplitude is good because the Zener diode and the base-emitter voltages are mutually compensated. Short switching times of the output transistors (10 μs) are obtained by high current driving the optocoupler diode (50 mA). Both simulation and measurements have shown that the output stage produces pulses with the amplitude range from 0 to 140 mA with supply less than 150 V voltage, frequency f = 5–100 Hz, and pulse duration t = 10–500 μs. The amplitude accuracy is in range of $\pm 10\%$ when the muscle resistance R_P varies between 0 and 1 kΩ. If R_P is constant, the accuracy is better than 1%. Time control is provided from a quartz crystal oscillator, hence it is very precise.

B. The Power Source

The battery-operated power supply unit generates galvanically isolated high voltages (150 V), two per output stage. The power supply is realized as a DC/DC converter to obtain high efficiency, small size, and light weight. The current programmed flyback configuration is adopted as the best solution for a DC/DC converter. Simplicity and low cost are the advantages of the flyback topology. Current programming with constant space and variable operating frequency provide a simple converter control circuit, stable control loop, and reliable overload protection [17].

The block diagram of the desired converter is shown in Fig. 4(a). The oscillator generates a square wave voltage, V_B, with variable frequency, $f = 1/T = 1\ (t_{\mathrm{ON}} + t_{\mathrm{OFF}})$, where t_{ON} is a variable pulse time and t_{OFF} is the constant space time duration. The maximum pulse time t_{ONMAX} is internally limited to protect the DC–DC converter against overload. The power switch is driven by the oscillator output to load energy into the transformer. The multiple output power transformer and rectifiers with a simple capacitive filter, C, provide tracking of output voltages and control voltage, V_C, which is used in the feedback loop after attenuation by the resistive divider $R - \alpha R$. In this manner, multiple galvanically isolated output voltages are available with a simple control circuit.

The DC/DC converter is shown in Fig. 4(b). Due to the low battery voltage, a control circuit is realized by single-supply operational amplifiers and comparators. Comparator U1/B and its passive elements form an oscillator which generates a square wave voltage with constant pulse and variable space duration, depending on the output voltage of comparator U1/A. The oscillator output controls driver transistors Q1, Q2 of the output power switch Q3. The driver energy consumption is low because transistors Q1, Q2 can not conduct simultaneously. The diode, D2, speeds up the MOSFET's turn off. To drive power switch Q3 with sufficient amplitude, auxiliary trans-



R1,R2,R4,R13 - 150K
R3 - 33K
R5,R6 - 82K
R7,R16 - 33K
R8,R11 - 10K
R01,...RO8 - 910K
R12,R14 - 3.9K
R15 - 62K

C1 - 120pF
C2 - 100nF
C4 - 100uF
CO1,...CO8 - 10uF
C6 - 1nF
C7 - 220pF

D1,D2,D7,D8 - 1N4148
D3,D4 - 1N5817
DO1,...DO8 - MUR140
DZ1 - LM385
Q1,Q2 - BC337
Q3 - VNE011
U1 - LM393
U2 - LM358

TRANSFORMER DATA:
CORE: RM10, N41, SIEMENS
AIRGAP: lp=0.15mm
WINDINGS: primary 12 tourns
1x8V secondary 20 tourns
1x5V secondary 16 tourns, bifilary
8x150V secondary 360 tourns

(b)

Fig. 4. The block diagram (a) and the electric scheme (b) of the DC–DC converter.

former winding L2, rectifier D3, and capacitive filter C2 are added. The MOSFET transistor's pinchoff must be less than the battery voltage to ensure starting of the converter operation.

The transformer, rectifiers, and simple capacitive filters are standard realizations. The slightly attenuated control voltage, V_C, is subtracted from the reference voltage V_R by the error amplifier U2/B, which is compensated by the differentiating circuit R13, C7. The level shifter U2/A increases the current sensing voltage obtained from the output transistor; this provides effective comparisons of the error amplifier's output and low value of sensing voltage (Fig. 4(b)). The channel resistance of the conducting MOSFET is used as a current sensing resistance to decrease converter dissipation. The diode D7 and capacitor C6 diminish the influence of the transistor switching spikes on the control circuit. The diode D8 breaks off the high drain voltage from the level shifter input in the process of turning off the MOSFET.

The converter is designed for a battery supply in the range from 3 V to 6 V, and eight 150 V/2 mA outputs. SPICE simulations showed high converter efficiency (75%). Laboratory tests of designed circuit confirmed simulation results. Converter standby consumption was under 30 mW. The output control voltage was obtained within $\pm 1\%$ accuracy. Other tracking output voltages remain within $\pm 10\%$, depending on the load distribution. Overload protection for all outputs was realized on the primary side of the transformer. The proposed circuit is stable under all operating conditions. The converter is inexpensive and has small volume (< 20 cm^3) and low weight.

The stimulator consumes about 1.5 W of power from a 6-V battery at medium output pulse amplitude (70 mA), pulse rate (50 Hz), and pulse duration (250 ms) when all four channels are active. The rechargeable Sony NP-60D battery with a 1-Ah capacity can drive the stimulator under the described conditions for about four hours continuously without recharging.

C. The Control Unit

The basic function of the control unit is to drive the output stages with the desired pulse duration and frequency. In addition, the control unit must provide a powerful interface with the user, a host computer, and other stimulators and sensors.

The control unit is shown in Fig. 5. The 68HC11 microcontroller is the core of the device [13]. The microcontroller operation is supported by the reset and supply voltage monitoring circuit TL7705 and by the RS232 converter MAX232, which provides standard voltage levels for asynchronous serial communication. The DIP switch is used to change the operation mode of the 68HC11 microcontroller. The bootstrap mode

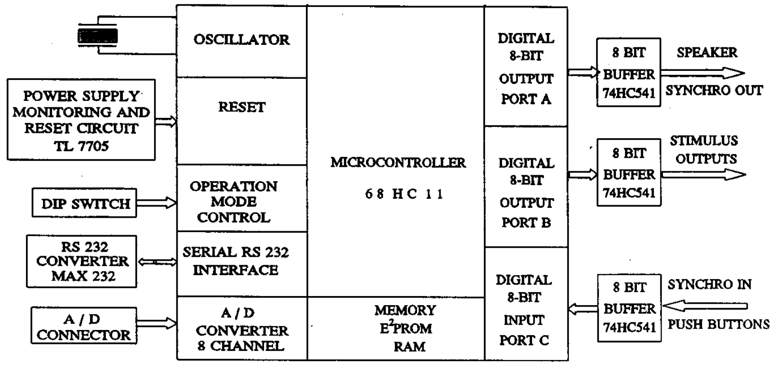


Fig. 5. The scheme of the control unit based on the 68HC11 microcontroller.

is used to download programs into the E^{2}PROM memory of microcontroller from the host computer by the serial RS232 interface. The single-chip mode microcontroller operation is used when the stimulator operates as an autonomous device. The A/D converter inputs of microcontroller can accept eight analog voltages within the range of 0 to 5 V. The output stages, the loudspeaker, and the synchronization output are driven by digital output ports trough buffers 74HC541. The synchronization input and push-buttons are buffered and connected to the digital input port of the microcontroller. The subroutine shown in Fig. 6(a) is used for generating pulses which drive the output stages of the stimulator. The pulse duration is controlled by software loops. Therefore, pulses at different outputs can not be generated simultaneously; they appear in sequence. There are various ways to provoke the generation of pulses: 1) periodic pulse generation at the desired frequency f is possible using a timer with four output compare registers, one for each output channel, as shown in Fig. 6(b); 2) the input synchronization pulse disables the periodic output pulse generation and generates output pulses by calling the subroutine described in Fig. 6(a); and 3) in sensor driven applications, both the output pulse frequency and the pulse duration are controlled by sensor signals and an application program.

An effective interface between stimulator and user is achieved through push-buttons (command signals) and a loudspeaker (feedback). The host computer interface is realized by serial RS232 interface. Synchronizing inputs and outputs are used to connect several stimulators. In such a case, the stimulator delivering synchronization pulses works as a master unit, while the others operate in the slave mode. The A/D inputs serves as an interface to be used for closed-loop control. The stimulator at this stage can accommodate up to eight analog and eight digital inputs.

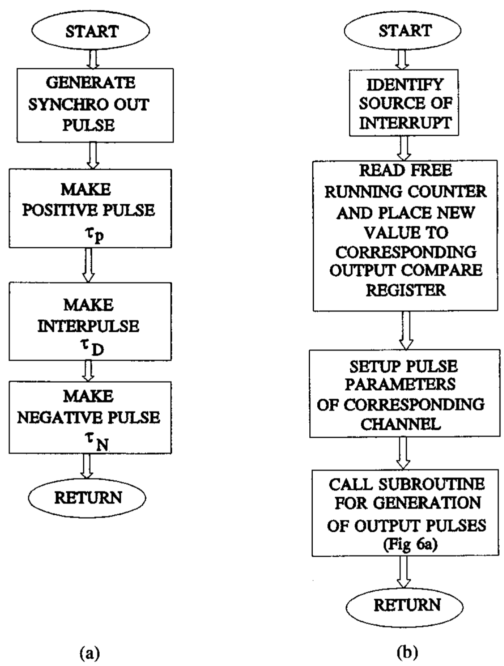


Fig. 6. The flow charts for subroutine which: (a) generates output pulses and (b) defines cyclic (periodic) repetition of the output pulses.

TABLE I
THE RANGE AND RESOLUTION OF THE STIMULATION PARAMETERS

	Minimum	Maximum	Resolution
Pulse rate [Hz]	5	99	2
Pulse duration [μs]	10	500	10
Amplitude [mA]	0	140	2

III. Applications of the Programmable Stimulator

The size of the stimulator is 18 cm × 10 cm × 3 cm and the mass $m = 320$ g. The front panel has four push-button switches: selection of the program, selection of channels, adjustment of the pulse duration, and rate. A Lithium rechargeable battery is attached to the box using a standard battery holder and it can by easily replaced. The stimulator has connectors for: 1) four pairs of electrodes, 2) synchronization in/out pulses, 3) RS232, and 4) sensors. Resolution and range of parameters are summarized in Table I.

The stimulator operates in Setup (SM) and Autonomous Modes (Fig. 1). In the SM, the user-friendly software for the host computer provides easy programming of the stimulator.

Additional software serves to download created programs to the stimulator using RS 232 protocol. In the AM of operation, the stimulator generates output pulses in three different ways: preprogrammed cyclic operation, controlled operation by an external synchronization input, and controlled operation by signals from sensors allowing sensory driven control of the stimulation [15]. In all cases, the synchronization pulse appears before output pulses. This pulse can be used to synchronize the operation of several stimulators and for the control of the stimulation artifact blanking circuit if natural sensors (e.g., signals from nerve cuffs or electrical activity of stimulated muscles) are to be used within the closed-loop control.

The cyclic stimulation pattern is meant for exercise and training routines when several muscles groups are stimulated for a fixed period of time in a preprogrammed sequence.

In the SM, the therapist or other nonexpert can set the parameters or stimulation patterns by using a Windows-based interface. The PC host computer displays the range of variables available for change. The program allows changes of the frequency of stimulation, pulse duration, rise and fall times, duration of the train of pulses in the exercise regime, timing between pulse delivery on different channels, and accommodates different preprogrammed sensory driven or closed-loop control systems (e.g., artificial reflex control [15]).

IV. Conclusion

A portable, programmable, battery-operated stimulator was designed for FES systems with the following features: 1) four channels; 2) bipolar, compensated monophasic stimulation pulses; and 3) controllable stimulation pattern. The stimulator has been tested for the restoration of standing and walking. It works according to the stated specifications. A series of stimulators is being prepared for clinical testing. The most important technical features of the stimulator are low power consumption, comfort, and simple programming with push-buttons and auditory feedback. The flexibility in programming and the possibility of using several stimulators in a master–slave configuration is another important technical characteristic. The synchronization pulse allows the integration of the stimulation artifact blanking circuit for eventual use with recordings from peripheral nerves or stimulated muscles.

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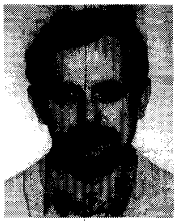
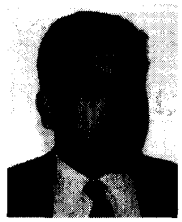
Milan Ilić was born in Čačak, Yugoslavia, on November 15, 1960. He received the B.S. and M.S. degrees in electrical engineering from the Faculty of Electrical Engineering, Belgrade, Yugoslavia, in 1984 and 1987, respectively.

He is presently a Teaching Assistant in the Department of Electronics at the Faculty of Electrical Engineering, Belgrade. His interests include analog and digital electronics, and control.



Dragan Vasiljević was born in Tabanovići, Yugoslavia, on September 10, 1950. He received the B.S., M.S., and Ph.D. degrees in electrical engineering from the University of Belgrade, Belgrade, Yugoslavia, in 1974, 1977, and 1984, respectively.

Since 1975, he has been with the Faculty of Electrical Engineering, University of Belgrade, where he is now an Associate Professor in Electronics. His current research interests are in the area of digital system design.



Dejan B. Popović was born in 1950. He received the B.S., M.S., and Ph.D. degrees from the Faculty of Electrical Engineering, University of Belgrade, Belgrade, Yugoslavia, in 1974, 1977, and 1981, respectively.

He is a Professor of Biomedical Engineering at the University of Miami College of Engineering, and at the Faculty of Electrical Engineering, University of Belgrade, Belgrade, Yugoslavia. He is an Adjunct Professor at the Division of Neuroscience, University of Alberta, Edmonton, Canada. He is the Senior Fellow at the Miami Project, University of Miami, Miami, FL. His scientific interests are in better understanding of biological organization of motor functions in humans and animals, and employment of these findings in designing assistive systems to improve the mobility of humans with physical disabilities. He is the author and coauthor of several textbooks and numerous scientific publication in journals, books, and conference proceedings.