Design and Development of Functional Electrical Stimulation Circuit

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Functional electrical stimulator (FES) is a device that creates controlled electrical pulses with a specific waveform to contract or relax muscles when it's placed in the correct position. The device is useful for patients with strokes or any spinal injury and they have lost or weakened the nerve control to their muscles. This study focuses on the design and development of a portable, single channel, biphasic and inexpensive FES circuit using Arduino Pro Mini [1]. A 7.4 V two-cell Li-ion battery feeds a 20 kHz PWM-driven DC–DC boost converter (L = 100 µH, C = 100 µF) that supplies up to 40 V to a MOSFET H-bridge. A resistive divider closes the voltage loop and holds the feedback signal within ±0.1 V of its set-point, while a 10 Ω shunt and an LM358 amplifier (28.5 mV/mA) monitor the load current. If the output reaches 40 V, the device stops stimulating for 2 s. Symmetric biphasic pulses were delivered at 30 Hz, with pulse widths of 224–1280 µs and a 150 µs interphase gap. The device operates entirely through the skin with no need of invasive methods using surface electrodes. The working frequency used is 30Hz, the pulse width is varied from 224-1280µs. Through experiments the output voltage was observed from 12.0-30.0 Vpp and PWM duty cycle from 19-62%. Beyond this level, the subjects began to experience mild discomfort. The experiments were performed on healthy subjects with the followed data. For participant 1 the voltage range was 12.0-22.6 Vpp and PWM duty cycle ranged from 19-49%. Respectively for participant 2 the voltage was 14.0-30.0 Vpp and PWM duty cycle from 23-62%. The reason for this variation can be different like ion concentration in muscles, sensitivity, skin thickness or subcutaneous fat layer. The FES system at the moment works as a voltage-controlled system but for future improvements multi-channel direct current controlled systems can be added. Functional electrical stimulation (FES) is a technique that uses low electrical currents to stimulate the nerves of the body muscles that has lost their senses such as in paralysis. This paper presents the design and development of a single-channel, biphasic FES device using an Arduino controller. The circuit consists of a DC-DC boost converter to generate high voltage, an H-bridge to produce biphasic pulses, a feedback mechanism for output control, and a pulse width modulation module. The device stimulates wrist flexion through surface electrodes on the Flexor Digitorum Superficialis (FDS) muscle. Experiments conducted on two healthy subjects demonstrate the ability to control pulse width (224 − 1280µs) and the voltage amplitude (up to 30V) to elicit muscle contractions. Mild wrist flexion is observed above 12 V. The portable Arduino-based device offers advantages such as cost-effectiveness, precise output regulation through feedback, and safety enhancements to prevent reverse current. Further clinical testing can affirm the therapeutic applicability for neuromuscular conditions. This FES device system can be enhanced to multi-channel capabilities and adaptive control algorithms.

Keywords: Functional Electrical Stimulation; Biphasic Pulse; Boost Converter; Closed-Loop Voltage Control; Portable Stimulator; Wrist Flexion; Rehabilitation.

Abbreviations:

ADC	Analog-to-Digital Converter
CCM/DCM	Continuous/Discontinuous Conduction Mode
CV	Coefficient of Variation
FDS	Flexor Digitorum Superficialis
FES	Functional Electrical Stimulation
NPRS	Numeric Pain Rating Scale
PWM	Pulse-Width Modulation
SD	Standard Deviation
Vpp	Peak-to-Peak Voltage

Introduction:

Functional electrical stimulation (FES) is one of the techniques that uses low electrical currents to stimulate the nerves supplying muscles in patients with conditions such as stroke, spinal cord injury, or multiple sclerosis. FES stimulates nerve fibers to trigger muscle contractions similar to natural activation by the central nervous system of our body, which allows patients to perform day-to-day functional tasks. FES devices generate short pulsed currents via surface or implanted electrodes by depolarizing motor neurons. Key technical parameters are pulse shape, amplitude, duration, and frequency. Biphasic symmetrical pulses with constant interphase intervals are preferred to ensure charge balancing [2]. Pulse frequencies of 10–60 Hz are typical for muscle stimulation.

This paper focuses on designing a single channel, biphasic, voltage-controlled FES device with closed-loop feedback and current monitoring for wrist and finger flexion using an Arduino as controller. The system consists of a DC–DC boost converter to generate high voltage from a 7.4 V two-cell (2S) Li-ion battery; an H-bridge circuit that generates biphasic pulses; voltage-divider feedback for closed-loop output-voltage regulation and shunt-resistor current monitoring; pulse width and amplitude modulation for the desired stimulus; and electrodes placed on the FDS muscle. The aim is to develop a low-cost, portable FES system with precise output regulation. Performance is evaluated by stimulating wrist flexion in healthy subjects. This can provide a foundation for multi-channel devices with improved muscle control. Potential applications include stroke rehabilitation and improvement of hand function.

There are two main types of stimulator in the market: the constant voltage type [3] and the constant current type [4][5][6][7]. The constant voltage type has an advantage in preventing discomfort with changes of load resistance due to electrode contact, but the stimulating current may decrease, which degrades stimulation performance. The constant current stimulator has more stable stimulating performance since a constant charge is supplied to the nerve regardless of impedance changes on the skin. This method, however, may cause discomfort due to the voltage increase needed to overcome changes in load impedance. Prior works have explored current-controlled electrical stimulation systems [8][9][10][11]. While advancing FES, most are either single-channel or require multiple power supplies without integrated feedback regulation. There is a need for portable solutions which provide enhanced control and safety.

Further prior art in FES systems includes multi-channel designs and closed-loop control approaches as well. [12] developed a 24-channel biphasic stimulator which uses high-voltage ICs to support wrist and finger gestures; FPGA-based control avoided unipolar induced tissue damage. [13] presented a 4-channel IoT electro-stimulator built from boost converters, H-bridges and an inertial sensor, with PID control of the ankle joint at 200 μs and 50 Hz. For high-voltage biphasic pulse generation, Tamtrakarn proposed a dual-output boost converter topology using a single inductor and supply, generating pulses above 100 V [6]. [14] presented an easy-to-implement multimodal electro-stimulator based on a PIC16F873A microcontroller, a boost converter, an H-bridge and a power management unit; however, it has no safety features to prevent accidental or excessive stimulation. [15] developed a programmable FES system that enables grasping through wrist flexion using an op-amp-based current source and an accelerometer, but it has not been tested in a clinical setting.

Recent Developments (2022–2026):

Many of the stimulators published since 2022 are portable, open-source or cheap to build. Trout et al. used an Arduino and a bidirectional Howland current source to make a three-channel stimulator with ±150 V compliance and a rise time of about 2 μs [16]. [17] shared the design of an eight-channel constant-current stimulator for FES cycling that can give biphasic pulses of up to 150 mA, 1000 μs and 100 Hz. For faster rise times, Wang et al.

paired a flyback converter with a switched-capacitor output stage in a small battery-powered unit [18]. Multi-purpose stimulators aimed at clinics were reported by [19] and by [20].

Chip-level work has focused mostly on efficiency and on keeping the charge balanced. [21] reached 86% end-to-end efficiency at 6 mA by letting a boost converter supply only the voltage that the electrode actually needs. [22] held the residual charge below 3 mV using two control modes, and [23] lowered the compliance voltage needed by implementing pulse-frequency modulation technique. Electrodes are changing as well. In a textile sleeve driven at 40 Hz and 300 µs, [24] found that dry electrodes gave the same torque as hydrogel pads and were a little more comfortable.

On the control side, newer systems decide when and how hard to stimulate from surface EMG and inertial sensors [25][26], or from a brain–computer interface [27]. The clinical picture is encouraging but incomplete. For upper-limb rehabilitation after stroke, one review found Fugl-Meyer improvements of around 5.4-5.6 points with FES [28], and another found 7.8 points when FES was combined with robotics [29]. The same literature points out that devices are still hard to carry, calibrate and put on [30], and that there is little agreement on stimulation parameters [31].

Limitations:

Limitations of existing systems include lack of feedback mechanisms for precise output control [8][9][10][11], reliance on multiple power sources [9][11], and minimal safety features to prevent over-stimulation [8][9][10][11]. Most lack buffered feedback for output regulation and rely on bench-top equipment for control and monitoring. Where cheap designs do have feedback, the papers rarely give numbers showing how well the output is regulated, and fewer still test the device on people.

Research Objectives:

The objectives of this work are to:

Design and Development of a DC–DC boost converter that raises a 7.4 V 2S Li-ion battery to a regulated output of up to 40 V.

Regulate the output voltage in closed loop so that the feedback signal stays within ±0.1 V of the set-point by the user (about ±1.1 V at the output).

Generate symmetric biphasic pulses at 30 Hz, with pulse widths adjustable from 224 to 1280 µs and a 150 µs interphase gap to keep the frequency constant.

Monitor the load current through a 10 Ω shunt resistor (28.5 mV/mA after amplification) and stop the output for 2 s when it reaches 40 V or over current detected.

Test the device on two healthy participants and record the output voltage, PWM duty cycle and self-reported movement at 20 parameter settings, keeping discomfort below 7 on the Numeric Pain Rating Scale.

Novelty and Contribution:

Previously developed low-cost stimulators do not have output regulation and shut off functions [14], require multiple power sources [9][11] or use an untested current source [15]. More recent open-source designs have used a Howland Current Source to achieve ±150V compliance [16]. Other designs have increased the number of channels and added IoT components [13][17]. This design has four key differences:

A single 7.4V battery is used and there is no mains connection or additional power source.

Both phases are created using one boost converter and a discrete level-shifted H-bridge meaning the amplitudes of both phases are equal.

The closed-loop regulation, shunt current sensing and over-voltage trip are all controlled using 8-bit Arduino Pro Mini [1].

The output voltage and duty cycle are recorded for all 20 settings for 2 participants. Few low-cost designs have this level of detail on published results.

Materials and Methods:

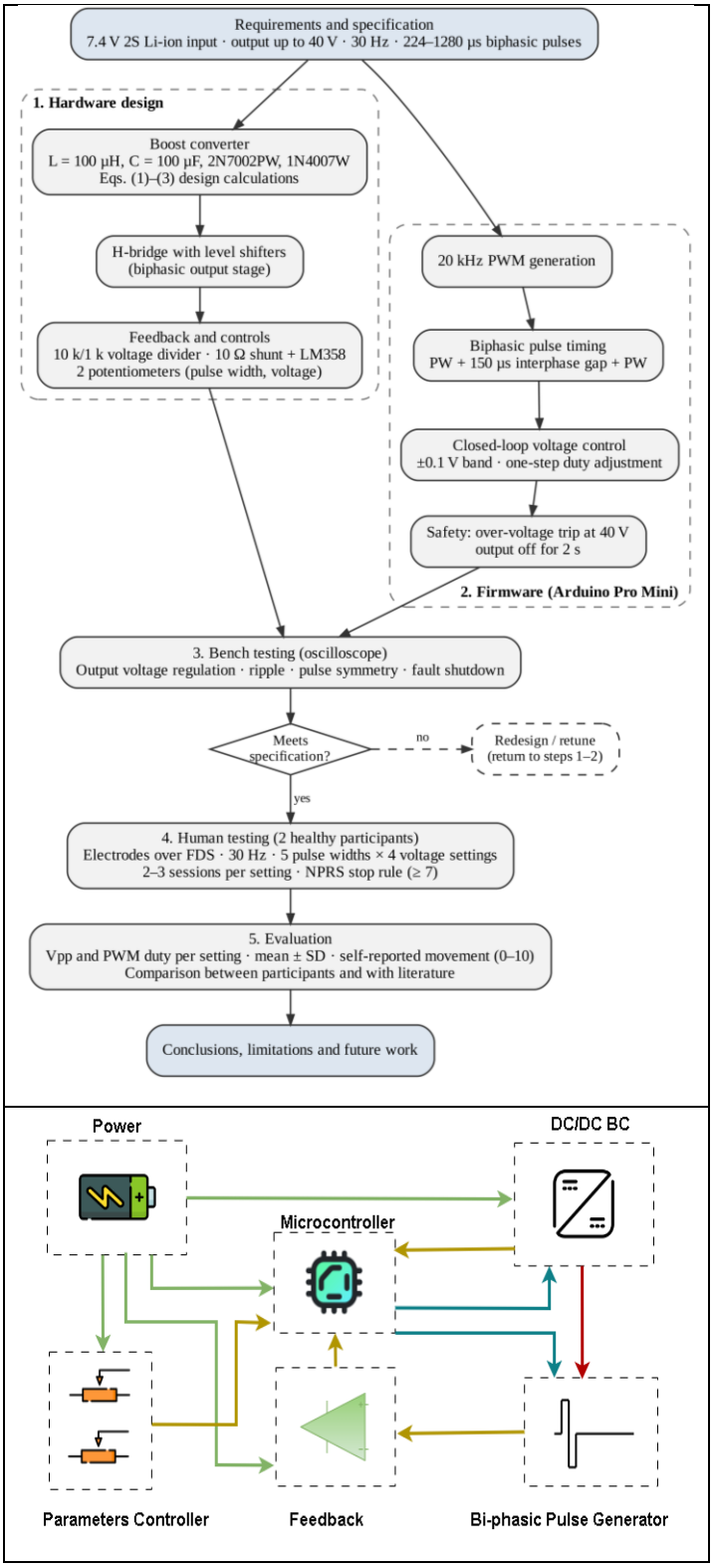


Figure 1. (a) Research flowchart of the study: hardware design, firmware development, bench testing, human testing and evaluation. (b) Functional diagram of the proposed system.

The materials and methods used to develop a practical electrical stimulation circuit are discussed in the section that follows. This project was funded by the National Research Program for Universities (NRPU), Higher Education Commission (HEC), Pakistan. The

research workflow is shown in Figure 1(a) and the functional diagram of the system in Figure 1(b). This is a closed-loop system and consists of a power supply, a microcontroller, a DC/DC voltage boost converter, and an H-Bridge, as illustrated in Figure 1. In this study, a Functional Electrical Stimulator was developed using an Arduino Pro Mini [1] microcontroller [14] for controlling the logic, feedback, and other relevant circuitry. The FES was designed to generate a biphasic waveform with variable duty cycle and voltage amplitude. The following materials were used in the development of this device: Arduino Pro Mini [1] microcontroller, DC/DC Boost converter [15], H-bridge [16], Potentiometer, Operational Amplifiers for feedback [17], Electrodes, and 7.4 V two-cell (2S) Li-ion battery. The microcontroller was programmed to generate a constrained Pulse Width Modulation (PWM) signal, which was then fed as an input to the boost converter. The boost converter generated high voltage from the 7.4 V two-cell (2S) Li-ion battery. The high voltage generated was then fed into an H-bridge circuit, which converted a DC signal to a biphasic feedback mechanism. The feedback signal was fed back into the microcontroller. A potentiometer was also used to control the pulse width of the stimulus. The electrodes were connected to the output of the H-bridge circuit, and the stimulus was applied to the targeted muscles of the person. To evaluate the effectiveness of the FES, a pilot study was conducted on two healthy adult participants.

Ethical Statement:

This study was performed in accordance with the Declaration of Helsinki. This study was performed in accordance with the Nuremberg Code. Ethical approval was not received for this human study because the device that was developed is of a non-interventional nature. The muscle stimulator was designed to produce mild stimulus current to arm muscles, that is in no way dangerous to the subject and is conducted by the developers on themselves, with full control on the current. All adult participants provided written informed consent to participate in this study.

Electrical Safety Measures:

The following measures were applied in all experiments:

- The firmware limited the boost output to 40 V. If the output reached 40 V, for example with a shorted or detached electrode, the PWM and H-bridge signals were switched off for 2s.
- The voltage set-point could not exceed the potentiometer maximum (ADC count 700), and the duty cycle changed by one step per control iteration, so the intensity could not jump suddenly.
- The two phases had the same amplitude and width and were separated by a 150 μs interphase gap, so no net DC current reached the skin [2].
- An LM358 amplifier buffered the shunt signal from the circuit, which kept the high-voltage stage away from the microcontroller input.
- Stimulation stopped at once if discomfort reached 7 on the Numeric Pain Rating Scale [32], or whenever a participant asked.

Hardware Design:

The hardware of the Functional electrical stimulator is designed with five modules, which contains a DC/DC boost converter, an H-bridge, a feedback circuit, a processing unit, and a parameters control circuit as shown in Figure 2. The processing unit is a microcontroller (Arduino Pro Mini [1]) to (1) generate the PWM signal for the high speed switching of the transistor in boost converter; (2) generate a timing signal to control the transistors of H-bridge which creates a biphasic waveform from the boosted DC voltage; (3) analyze the feedback coming from the Op amp to ensure a stable boosted voltage; (4) monitoring the manual parameters controlled by the Potentiometers; (5) detect an abnormal electrode condition, when the output reaches 40 V, and stop the stimulation for 2 s; (6) adjust the PWM.

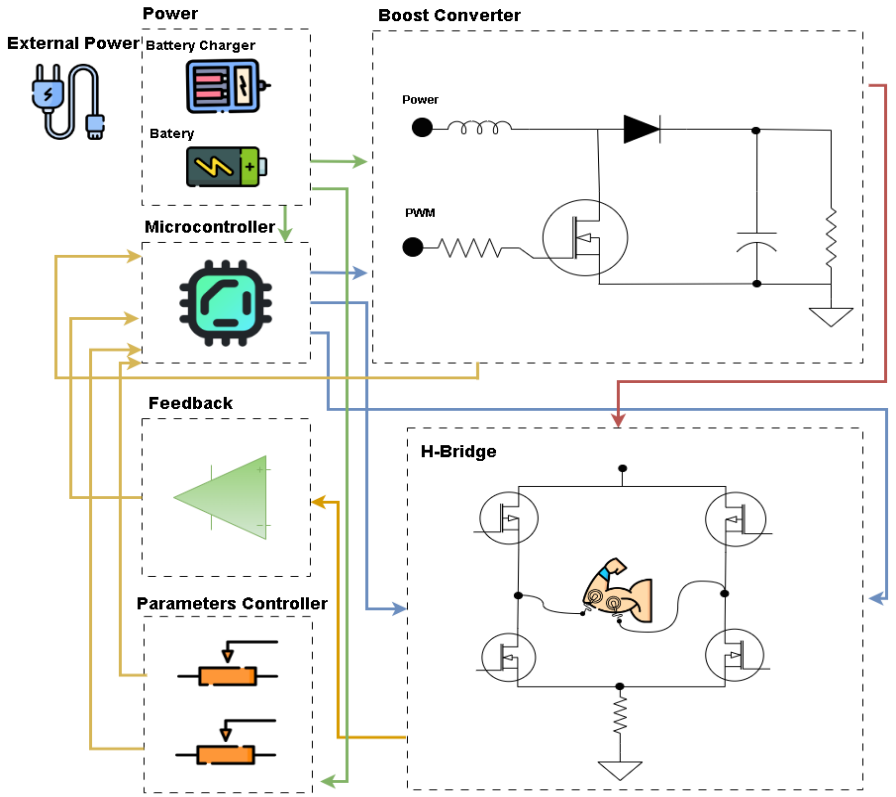


Figure 2. Hardware Architecture

The H-bridge [33] and boost converter [34] are crucial parts of the stimulator, and the circuits are presented in Figure 3. There are two parameters of the FES device which we are controlling using manual parameter controllers. In this part of the circuit, we are using two potentiometers to control the pulse width of the muscle stimulus signal and the boost converter voltage, and the circuit is presented in Figure 3. The analog outputs of this module are fed into the microcontroller for comparison and generation of required PWM and pulse width of the stimulus. The DC/DC boost converter module is boosting the 7.4 V two-cell (2S) Li-ion battery voltages. The pulse width modulated (PWM) signal is applied at the gate of MOSFET, as presented in Figure 3. To keep the required high voltage at the output of BC in mind, the value of the inductor placed in the BC can be schematized as shown in the Equation (1) [15][18] waveform by precisely controlling the switching of transistors in the H-bridge using the microcontroller.

The boost inductor was sized with the standard boost-converter relations [35][36]. In continuous conduction, the ideal duty cycle is

$$D = \frac{1-Vi}{Vo} \quad (1)$$

and the inductance needed for a given ripple current is

$$L = \frac{(Vi \cdot D)}{(fs \cdot \Delta IL)} \quad (2)$$

The boundary between continuous and discontinuous conduction is

$$L_{crit} = \frac{D\,(1-D)^2\,RL}{(2\,fs)} \quad (3)$$

where L is the inductance (H), Vi the battery voltage (V), Vo the boost output voltage (V), D the duty cycle, fs the switching frequency (Hz), ΔIL the peak-to-peak inductor ripple current (A) and RL the equivalent load resistance (Ω).

The H-bridge applies +Vo and −Vo across the electrodes, so the peak-to-peak stimulus is 2Vo. With Vi = 7.4 V and fs = 20 kHz, Eq. (1) gives D = 0.51 at Vo = 15 V (30 V_{pp}, the highest level used on participants) and D = 0.82 at the 40 V maximum. The stimulator

draws only a few milliamperes on average; the 11 kΩ feedback divider alone draws 1.4 mA at 15 V. RL is therefore several kilohms, and Eq. (3) puts L_{crit} in the millihenry range.

The selected 100 µH inductor (CD127R, 12 × 12 × 7 mm) is well below L_{crit}, so the converter runs in discontinuous conduction mode. We chose this on purpose. The inductor current returns to zero every cycle, the right-half-plane zero of the boost converter moves far above the loop bandwidth, and the firmware's step-wise duty control stays stable. The 100 µF output capacitor C1 holds the output between pulses. The measured duty cycles (19–62%) are higher than the ideal values because of losses in the 1N4007W diode, the MOSFET on-resistance and the inductor winding.

The feedback circuit, which is another important module of the FES device circuit, is based on the shunt resistance [37]. The output voltage is sensed by a 10 kΩ/1 kΩ divider (ratio 1/11), so 40 V gives 3.64 V at the ADC input. The load current flows through a 10 Ω shunt (R5), and the voltage across it is amplified by an LM358 in non-inverting configuration with a gain of 1 + R14/R12 = 1 + 10 kΩ/5.4 kΩ = 2.85. This gives a sensitivity of 28.5 mV per mA. The current flowing through the shunt resistor generates the voltages which are passed through an operational amplifier and fed into the microcontroller. The microcontroller processes the analog feedback and generates the stimulus and PWM accordingly. The use of operational amplifier also adds a layer of protection between high voltage generated at BC and applied to the skin and the microcontroller.

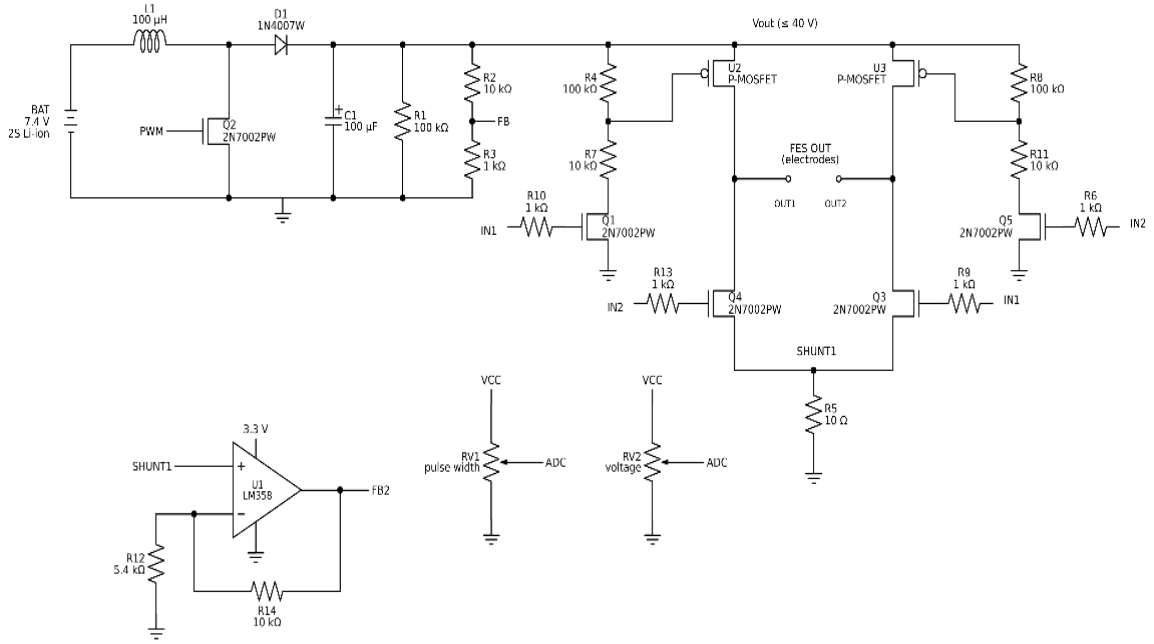


Figure 3. Functional Electrical Stimulator Circuit Schematics

Voltage-Controlled versus Constant-Current Stimulation:

A constant current stimulator forces a set current through the tissue, and its output voltage rises or falls with the skin–electrode impedance. A voltage-controlled stimulator fixes the voltage, and the current (I = V/Z) changes with impedance [3][4]. The proposed device is voltage-controlled. The closed loop regulates the boost output voltage, and the shunt current is measured only for monitoring and fault detection. Current is not the control variable, and the device does not claim constant-current regulation.

Table 1. Component Specification and Selection Criteria.

Block	Ref.	Component / value	Selection criterion
Power	–	2S Li-ion battery, 7.4 V (6.0–8.4 V)	Single supply, portable, no mains
Controller	–	Arduino Pro Mini [1] (ATmega328P, 10-bit ADC) [5 V/16 MHz or 3.3 V/8 MHz]	Low cost, small, hardware PWM timers
Boost	L1	100 µH, CD127R (12 × 12 × 7 mm)	DCM at light load (Eq. 3), compact
Boost	Q2	2N7002PW N-MOSFET, 60 V	Rated above the 40 V output, logic-level gate
Boost	D1	1N4007W rectifier, 1000 V	Voltage margin, low cost
Boost	C1	100 µF electrolytic	Holds output between pulses
Boost	R1	100 kΩ	Discharges C1 at switch-off
Feedback	R2/R3	10 kΩ / 1 kΩ (ratio 1/11)	40 V maps to 3.64 V, inside ADC range
H-bridge	Q3, Q4	2N7002PW N-MOSFET (low side)	60 V rating
H-bridge	U2, U3	P-MOSFET (high side) [P-MOSFET part no.]	Switches the high-voltage rail
H-bridge	Q1, Q5	2N7002PW level shifters; R4/R8 100 kΩ, R7/R11 10 kΩ	Drive high-side gates from logic level
H-bridge	R6, R9, R10, R13	1 kΩ gate resistors	Limit gate current
Current sense	R5	10 Ω shunt	0.1 V at 10 mA, low loss
Current sense	U1	LM358, gain 2.85 (R14 10 kΩ, R12 5.4 kΩ)	28.5 mV/mA, single supply
Controls	–	Two potentiometers (pulse width, voltage)	Read by 10-bit ADC
Electrodes	–	Surface electrodes [electrode pad type/size] with FES electrode cable (RehaMove)	Reusable, standard FES connector

Firmware Design:

The firmware was written using C++ and the Arduino framework. It had two loops. One for stimulation of the biphasic pulses and one for the control of the voltage by changing the duty cycle of the boost PWM. The firmware follows these steps:

Configure the I/O pins and start the boost PWM at 20 kHz with 0% duty cycle. Both IN1 and IN2 of the H bridge will be pulled low.

Read the duty cycle potentiometer and voltages. Use the ADC to convert them to values between 0 and 1023 (10-bit). Duty cycle will be scaled to give values between 224-1280 µs. Voltage will be limited to 700.

Calculate expected feedback voltage from the voltage divider. This can be calculated as VFB = Vset R3/(R2+R3). The voltage will be accepted within ±0.1V. The duty cycle will be increased by one if the voltage is below this range. Decreased by one if the voltage is above the range. This will continue until the voltage is within range. Since the duty cycle only changes by one at a time the output voltage will increase/decrease gradually.

Set IN1 high for one pulse width. Turn both IN1 and IN2 off f for 150µs. Set IN2 high for one pulse width. Turn both IN1 and IN2 off for the remainder of the 33.3ms (30Hz).

This means IN1 and IN2 are never on simultaneously. So there is no shoot through current in the H bridge.

If the output voltage reaches 40V (electrode shorted or disconnected) the boost PWM and H bridge signals will be turned off for 2 seconds and then go back to step 2.

Experimental Setup and Protocol:

This section of study discusses the design of test to evaluate the applicability and design of the functional electrical stimulator. An oscilloscope was used to analyze the working of H-Bridge and DC/DC Boost Converter modules. The factors evaluated were evaluated are (1) the ability of the designed stimulator to produce stimulus in the wrist muscles; (2) The effectiveness of the boost converter by analyzing the PWM value and the amplified voltages; (3) The effect of changing the pulse width of the stimulus signal.

The logic of the conducted tests was to analyze the behavior and stability of the circuit by varying the pulse width of stimulus, the variation in the applied voltage in the boost converter, and when the subject of the test is changed. The tests were conducted on multiple healthy persons by applying the Biphasic stimulus. The frequency of the stimulus signal was 30 Hz, and it was kept same for all the tests.

The tests were performed on two healthy subjects (male 1: 1.68m, 62kg, Body Mass Index 22kg/m^2 and male 2: 1.68m, 60kg, Body Mass Index 21.3kg/m^2). The parameters of electro-muscle stimulator were the following: varied pulse duration ranging from minimum 224μs to maximum 1280μs, frequency of 30Hz, and variable amplitude (duty cycle) spanned from 19% to 62%. The electrodes were placed over the Flexor Digitorum Superficialis muscle, and the position of the electrodes are shown in Figure 4.

Two healthy adult males were chosen because this is a proof-of-concept study of the device's electrical performance and safety, not of therapeutic effect. Healthy participants give an intact, repeatable neuromuscular response, so changes in output can be attributed to the device and not to impairment, and no patient was exposed to an unvalidated device.

Surface electrodes were connected through a FES electrode cable and placed over the FDS muscle belly, 3 inches (7.6 cm) apart (Figure 4). Each of the 20 settings (five pulse widths at four voltage settings) was applied in 2–3 sessions of 10–15 min each. The peak-to-peak output voltage across the electrodes and the PWM duty cycle were measured with an oscilloscope. After each setting, the participant rated the wrist and finger movement on a 0–10 scale (0 for no movement, 10 for full wrist flexion) and the discomfort on the NPRS. The values in Tables 2 and 3 are one representative session / mean of the 2–3 sessions. Repeatability was assessed from the spread of the output across the five pulse widths at each set-point (Table 4).

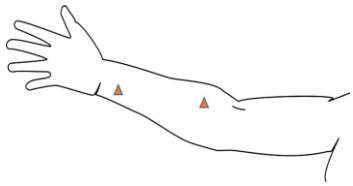


Figure 4. Electrode placement over the flexor digitorum superficialis on the anterior forearm

The parameters for stimulation were determined so that the individuals' safety is ensured along with the enhancement of stimulation signal and to minimize the participant discomfort. Discomfort or pain levels were assessed using the Numeric Pain Rating Scale throughout the course of the experiment. Participants were given clear instructions on how to use the scale, and they were asked to promptly communicate any changes they perceived [32]. The scale's range spans from 0 to 10, where 0 indicates the absence of pain or discomfort, and 10 represents the most extreme pain imaginable. It had been predetermined that if the reported pain or discomfort reached a level of 7 or higher, the experiment would be halted immediately,

or it could be stopped at the participant's request at any point. The subjects were sitting on a chair with both arms resting on the desk in these experiments. The individuals were instructed to relax the muscles and leave the arms in neutral position. The biphasic muscle stimulator signal was applied to produce wrist and finger flexion.

Results and Discussion:

This section reports the bench performance of the hardware and the results of the tests on two participants.

Hardware:

The hardware was constructed according to the design, featuring biphasic FES, a maximum output of ±40 V from the 7.4 V 2S Li-ion battery, a buffered feedback circuit, and potentiometer based 2 controllers for pulse width and applied voltage. Refer to Figure 5 for with buffered feedback.

As depicted in Figure 5, the core of the circuit is the processing module. It facilitates the generation of PWM, generation of signals to create a biphasic stimulus, and monitoring of the feedback from stimulus.

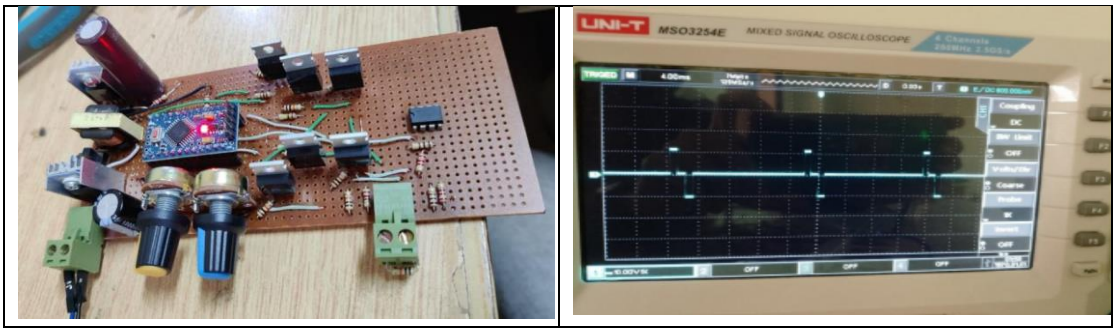


Figure 5(a). Constructed Electrostimulation Circuit: Illustrated Design of the FES device Circuit. On the left is the boost converter, in the center the processing module, and on the right, the H-Bridge module with isolated feedback. **(b);** it is the best evidence for symmetry.

Bench Performance:

Bench performance was measured with an oscilloscope. The boost converter held its set-point up to 40 V. Above 40 V the output fell under load, so 40 V is the maximum regulated output. Participants were stimulated at up to 30 V_{pp} (a 15 V boost output), mostly near 20 V_{pp}. The output ripple was below the oscilloscope's resolution at the range used and did not show on the trace.

The positive and negative phases had the same amplitude and width. Both phases are taken from the same rail with identical timing and a 150 μs gap between them, so the net charge per pulse is nominally zero. At a fixed set-point, the coefficient of variation of the output across the five pulse widths was 2.3–5.9% for participant 1 and 4.4–7.6% for participant 2 (Table 4). Converter efficiency and current stability were not measured in this study.

Physical Hardware Evaluation:

Table 2 and Table 3 present the results of the physical hardware test on two subjects. During the evaluation of the physical hardware, the frequency of the stimulus was kept constant at 30 Hz. The response of the stimulus signal and the values of different parameters were measured. The controllable values are the pulse width of the stimulus signal and the intensity of the signal which is peak-to-peak voltages. Pulse width varied across settings, while adjusting the stimulus intensity via the potentiometer. This analysis aimed to examine muscle stimulation and observe targeted muscle movement. After setting the value of the pulse width, the voltage value was controllable. The value of the second column is the output of potentiometer value sampled by 10-bit built-in ADC of the Arduino Pro Mini [1]. The 700 value is mapped to the maximum value in the firmware.

Table 2. Outcome of experiments conducted on participant 1

Pulse Width	Voltage Controller *	Voltage$_{p-p}$	PWM
224 µs	400	13 V	21%
	500	17.4 V	31%
	600	19.2 V	39%
	700	22.6 V	49%
512 µs	400	12 V	19%
	500	16 V	27%
	600	18.4 V	35%
	700	20 V	47%
768 µs	400	13 V	21%
	500	17.4 V	31%
	600	19.2 V	39%
	700	22.6 V	49%
1024 µs	400	12 V	20%
	500	15.2 V	27%
	600	19.6 V	39%
	700	21 V	47%
1280 µs	400	12.6 V	21%
	500	16 V	27%
	600	19 V	39%
	700	21 V	45%

Table 3. Outcome of experiments conducted on participant 2.

Pulse Width	Voltage Controller *	Voltage$_{p-p}$	PWM
224 µs	400	14 V	24%
	500	18 V	31%
	600	21 V	42%
	700	30 V	61%
512 µs	400	15.2 V	24%
	500	19.2 V	35%
	600	22.4 V	50%
	700	28 V	60%
768 µs	400	15.2 V	27%
	500	22 V	45%
	600	24 V	51%
	700	30 V	62%
1024 µs	400	15.6 V	26%
	500	19.2 V	33%
	600	21 V	45%
	700	28 V	58%
1280 µs	400	14.4 V	25%
	500	19.2 V	35%
	600	20.8 V	46%
	700	27 V	62%

Table 4. Output at each controller setting: mean ± SD across the five pulse widths

Controller	P1 Vpp (V)	P1 CV	P1 PWM (%)	P2 Vpp (V)	P2 CV	P2 PWM (%)	P2 − P1
400	12.5 ± 0.5	4.0%	20.4 ± 0.9	14.9 ± 0.7	4.4%	25.2 ± 1.3	+2.4 V (+18.8%)
500	16.4 ± 1.0	5.9%	28.6 ± 2.2	19.5 ± 1.5	7.6%	35.8 ± 5.4	+3.1 V (+19.0%)
600	19.1 ± 0.4	2.3%	38.2 ± 1.8	21.8 ± 1.4	6.3%	46.8 ± 3.7	+2.8 V (+14.5%)
700	21.4 ± 1.1	5.3%	47.4 ± 1.7	28.6 ± 1.3	4.7%	60.6 ± 1.7	+7.2 V (+33.4%)
All	17.4	–	–	21.2	–	–	+3.9 V (+22.2%)

With two participants, no inferential test is meaningful, so the results are summarized descriptively in Table 4

Across all tested pulse width values in Table 2, muscle movement was absent in subject 1 when the voltage controller remained below 500. A slight movement was however observed at 500 and 600 value of voltage controller for up to 1024 μs pulse width. A mild movement was observed for all values of pulse width when the voltage controller value was maximum. For the 1280 μs pulse width signal, mild movement was observed even at a voltage controller value of 600. Figure 6 illustrates a graphical comparison of muscle response and applied voltage. The data of the tests performed on second participant are presented in Table 2I. The performance of the developed FES exhibited a pattern for the second healthy subject that was very similar to that observed on the first subject, mentioned above. Figure 6 illustrates a graphical comparison of muscle response and applied voltage of the two participants.

The electrical muscle stimulation (EMS) device we designed and developed was based on the use of Arduino Pro Mini [1] as a controller. Unlike previous efforts which focused on applications or hardware components [13][14], this study introduced the complete hardware of a biphasic single channel FES device. The device was able to generate constrained PWM, which was fed as input to the boost converter, resulting in the generation of high voltage from the 7.4 V DC battery. This high voltage was then fed into an H-bridge, which created a biphasic stimulus. We also incorporated a feedback mechanism into the design to enable better control over the amount of voltage applied to the muscles of the person. In addition, we used a potentiometer to control the pulse width of stimulus and enhance the control over the output.

The device met its design targets within the limits set for participant safety. Like most low-cost stimulators [13][14], it uses an H-bridge, and it adds a buffered feedback path, an over-voltage trip at 40 V and a firmware limit on the set-point. Furthermore, safety measures were integrated into the firmware, restricting the output High Voltage to ensure subject safety.

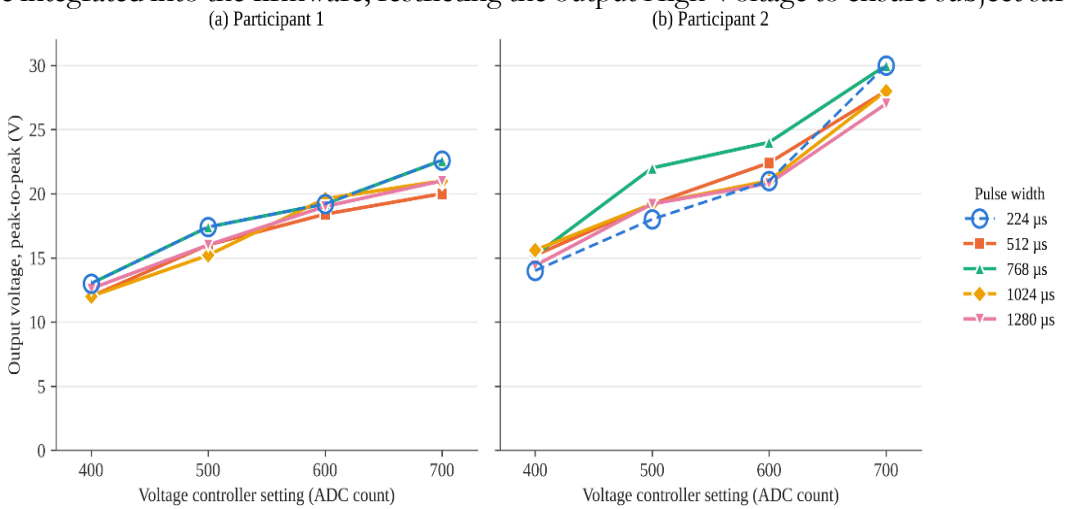


Figure 6. Effect of Applied Voltage on Muscle Stimulation-(left) Participant 1 Muscle Response to the Stimulus, (right) Participant 2 Muscle Response to the Stimulus

The FES device we developed has several advantages over conventional muscle stimulation devices. First, it is cost-effective, as the use of Arduino Pro Mini [1] as a controller and the other components used in the device are readily available and relatively inexpensive. Second, it is portable and runs from a single battery. Third, it is non-invasive and uses surface electrodes only. Its safety and usefulness for patients have not yet been tested.

When altering the pulse width of the stimulus signal, there's a lesser impact on hand and finger muscle movement compared to changes in applied signal intensity achieved by voltage adjustments. No hand and finger muscle movement were observed below 12 V. Movement started with voltages exceeding 12 V. Notably, Figure 6 illustrates an intriguing observation: lower pulse width yielded a steeper muscle movement intensity curve than higher pulse width. Higher pulse width led to more spontaneous movement variation in response to applied signal intensity changes.

Proof of Concept:

To demonstrate the feasibility of the proposed Arduino-based biphasic FES device, preliminary testing was conducted on two healthy adult subjects. The aim was to evaluate the ability to control key stimulation parameters and elicit targeted muscle contractions.

The Flexor Digitorum Superficialis (FDS) muscle was chosen for wrist flexion as it is a major flexor in the anterior forearm [38]. Bipolar surface electrodes were placed over the muscle belly as shown in Figure 4, with 3 inches (7.6 cm) inter-electrode distance.

Key Stimulation Parameters:

Pulse width and intensity, were systematically varied to characterize the FES response. Pulse width ranged from 224 μs to 1280 μs, adjusted via the microcontroller. Intensity was controlled by changing the output voltage of the DC-DC converter module from 12 V to 22.6 V. Frequency was fixed at 30 Hz.

As summarized in Tables 2 and 3, no visible muscle contraction occurred below 12 V for both subjects across all pulse widths. Above 12 V, mild FDS activation and wrist flexion was observed, with increased intensity at higher voltages. Consistent with Figure 6, lower pulse widths gave a steeper rise in movement once the threshold was passed. This indicates the ability to modulate movement through parameter adjustments.

The preliminary results demonstrate basic functioning of the FES device - it could deliver biphasic stimulation at the wrist flexor muscle in a controlled manner. The intensity and nature of muscle contraction could be varied by changing the stimulus voltage and pulse width respectively. This proof of concept on healthy individuals provides a foundation for efficacy testing in patient populations with neuromuscular deficits.

Limitations and Future Work:

The 30 Hz stimulation frequency is within the 10–60 Hz range typical for muscle stimulation [2] and below the 40 Hz and 50 Hz used by [24] and [13]. Our pulse widths (224–1280 μs) are longer than the 200–300 μs used in those studies and extend past the 1000 μs maximum of [17]. Visible wrist flexion began above 12 V_{pp} at every pulse width.

Most recent stimulators are current-controlled and report thresholds in milliamperes [16][17], so our voltage thresholds cannot be compared with theirs directly. Logging the shunt current would make that comparison possible. The 40 V limit is far below the ±150 V compliance of [16]. It was enough for the forearm, but larger muscles or high-impedance skin may need more.

Only two healthy male participants were tested. Movements were self-reported and not measured with any devices such as a goniometer or IMU, and current and efficiency were not logged. The results show that the device can produce wrist flexion in healthy forearms. The study has further limitations:

Movement was self-reported, not measured with a goniometer or IMU.

The shunt current was used only for fault detection; it was neither logged nor regulated.

Converter efficiency was not measured.

The 1N4007W is a standard-recovery rectifier; an ultrafast or Schottky diode would reduce switching losses at 20 kHz.

The device has not been tested against IEC 60601-2-10 [39].

An ESP32-based version with a 12-bit ADC and wireless control has been prototyped and will be evaluated in future work.

Conclusions:

This study designed and built a low-cost, portable, single-channel biphasic FES device controlled by an Arduino Pro Mini [1] and powered by one 7.4 V Li-ion battery. The boost converter regulated its output up to 40 V, and the firmware kept the feedback within ±0.1 V of its set-point. At fixed set-points, the coefficient of variation of the output across five pulse widths was 2.3–7.6%.

Symmetric biphasic pulses at 30 Hz and 224–1280 µs produced visible wrist flexion above 12 Vpp in both participants. The outputs were 12.0–22.6 Vpp for participant 1 and 14.0–30.0 Vpp for participant 2, at duty cycles of 19–62%, and participant 2 received on average 22% more voltage. Voltage had a stronger effect on movement than pulse width.

Once validated, the device could serve as a low-cost tool for hand rehabilitation post-stroke, in clinics and at homes, and as a platform for FES research and teaching. The present results are limited to two healthy participants and self-reported movement, without current or efficiency measurements.

Before any clinical use, we recommend testing the device against IEC 60601-2-10 [39] and adding current logging and a current limit to the firmware. Trials in stroke patients should then be run with ethics approval and objective outcomes such as the Fugl-Meyer upper-extremity score [28].

Declarations:

Authors contributions: All authors contributed equally to this project.

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Informed consent: Informed consent was obtained from both participants. Written informed consent to publish this paper was also obtained from both participants.

Data availability: The data supporting this study are given in Tables 2–4 and are available from the corresponding author on reasonable request.

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Conflicts of interest: The authors declare no conflict of interest.

References:

[1] "Arduino Pro Mini | Arduino Documentation." Accessed: Nov. 08, 2026. [Online]. Available: https://docs.arduino.cc/retired/boards/arduino-pro-mini/

[2] D. R. Merrill, M. Bikson, and J. G. R. Jefferys, "Electrical stimulation of excitable tissue: design of efficacious and safe protocols," *J. Neurosci. Methods*, vol. 141, no. 2, pp. 171–198, Feb. 2005, doi: 10.1016/J.JNEUMETH.2004.10.020.

[3] D. C. de Souza, M. do C. Gaiotto, G. N. Nogueira Neto, M. C. F. de Castro, and P. Nohama, "Power amplifier circuits for functional electrical stimulation systems," *Res. Biomed. Eng.*, vol. 33, no. 2, pp. 144–155, Jun. 2017, doi: 10.1590/2446-4740.07716.

[4] X. Li, S. Zhong, and J. Morizio, "16-Channel biphasic current-mode programmable charge balanced neural stimulation," *Biomed. Eng. Online*, vol. 16, no. 1, p. 104, Aug. 2017, doi: 10.1186/S12938-017-0385-0.

[5] C. C. Chen and K. T. Tang, "A 12V-500µA neuron stimulator with current calibration mechanism in 0.18µm standard CMOS process," *2011 IEEE Biomed.*

Circuits Syst. Conf. BioCAS 2011, pp. 57–60, 2011, doi: 10.1109/BIOCAS.2011.6107726.

[6] A. Tamtrakarn, Tamtrakarn, Atit, Tamtrakarn, and Atit, "A 115V bi-phasic pulse electrical muscle stimulator by using inductor-sharing dual-output boost converter with supply-stepping switch driver," *rcsy*, vol. 2018-January, p. 3, Jul. 2017, doi: 10.1109/ICECS.2017.8291997.

[7] K. W. E. Cheng, Y. Lu, K. Y. Tong, A. B. Rad, D. H. K. Chow, and D. Sutanto, "Development of a circuit for functional electrical stimulation," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 12, no. 1, pp. 43–47, Mar. 2004, doi: 10.1109/TNSRE.2003.819936.

[8] D. R. P. Chamal, M. I. M. Fernando, W. P. M. W. Kulathunga, K. D. Pathirana, and N. W. Prins, "Design of a Novel Current Controlling Module for Functional Electrical Stimulation (FES) System," *Proc. 6th Int. Conf. Inf. Technol. Res. Digit. Resil. Reinvention, ICITR 2021*, 2021, doi: 10.1109/ICITR54349.2021.9657272.

[9] A. Tamtrakarn, "A portable switching bi-phasic stimulator with level-shifting inverter for functional electrical stimulation operating under a 6V supply voltage," *IEEE Reg. 10 Annu. Int. Conf. Proceedings/TENCON*, vol. 2016-January, Jan. 2016, doi: 10.1109/TENCON.2015.7373081.

[10] H. P. Wang, Z. G. Wang, X. Y. Lu, Z. H. Huang, and Y. X. Zhou, "Design of a pulse-triggered four-channel functional electrical stimulator using complementary current source and time division multiplexing output method," *Proc. Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. EMBS*, vol. 2015-November, pp. 1671–1674, Nov. 2015, doi: 10.1109/EMBC.2015.7318697.

[11] B. Basumatary, R. S. Halder, and A. Sahani, "A Microcontroller based Charge Balanced Trapezoidal Stimulus Generator for FES System," *Conf. Rec. - IEEE Instrum. Meas. Technol. Conf.*, vol. 2021-May, May 2021, doi: 10.1109/I2MTC50364.2021.9459837.

[12] T. Ward *et al.*, "Multichannel Biphasic Muscle Stimulation System for Post Stroke Rehabilitation," *Electron. 2020, Vol. 9, Page 1156*, vol. 9, no. 7, p. 1156, Jul. 2020, doi: 10.3390/ELECTRONICS9071156.

[13] T. F. De Almeida, L. H. B. Borges, and A. F. O. de A. Dantas, "Development of an IoT Electrostimulator with Closed-Loop Control," *Sensors 2022, Vol. 22, Page 3551*, vol. 22, no. 9, p. 3551, May 2022, doi: 10.3390/S22093551.

[14] M. C. Santos *et al.*, "Easy-to-implement configurable multimodal electrostimulator," *INSCIT 2021 - 5th Int. Symp. Instrum. Syst. Circuits Transducers*, Aug. 2021, doi: 10.1109/INSCIT49950.2021.9557252.

[15] M. S. J. *et al.*, "A Functional Electrical Stimulator to Enable Grasping Through Wrist Flexion," *Int. J. Biol. Biomed. Eng.*, vol. 16, pp. 19–29, 2022, doi: 10.46300/91011.2022.16.4.

[16] M. A. Trout, A. T. Harrison, M. R. Brinton, and J. A. George, "A portable, programmable, multichannel stimulator with high compliance voltage for noninvasive neural stimulation of motor and sensory nerves in humans," *Sci. Rep.*, vol. 13, no. 1, Dec. 2023, doi: 10.1038/S41598-023-30545-8.

[17] T. Coelho-Magalhães, E. Fachin-Martins, A. Silva, C. A. Coste, and H. Resende-Martins, "Development of a High-Power Capacity Open Source Electrical Stimulation System to Enhance Research into FES-Assisted Devices: Validation of FES Cycling," *Sensors 2022, Vol. 22, Page 531*, vol. 22, no. 2, p. 531, Jan. 2022, doi: 10.3390/S22020531.

[18] H. Wang, R. Tang, and X. Liu, "Development of a Compact High-Voltage Functional Electrical Stimulation Device," Dec. 2024, Accessed: Nov. 08, 2026. [Online].

Available: https://arxiv.org/pdf/2412.12064

[19] R. Lakatem, S. Boontaklang, and C. Chompoo-inwai, "Advanced Design and Implementation of a 2-Channel, Multi-Functional Therapeutic Electrical Stimulator," *Electron. 2024, Vol. 13, Page 3793*, vol. 13, no. 19, p. 3793, Sep. 2024, doi: 10.3390/ELECTRONICS13193793.

[20] L. Popović-Maneski and S. Mateo, "MotiMove: Multi-purpose transcutaneous functional electrical stimulator," *Artif. Organs*, vol. 46, no. 10, pp. 1970–1979, Oct. 2022, doi: 10.1111/AOR.14379;PAGE:STRING:ARTICLE/CHAPTER.

[21] Y. You, R. Tian, Y. Zhang, Z. Chen, W. Lu, and Y. Zhang, "A High-Voltage-Compliant 86% Peak Efficiency Current-Mode Stimulator With Dynamic Voltage Supply for Implantable Medical Devices," *IEEE J. Solid-State Circuits*, vol. 60, no. 7, pp. 2606–2618, 2025, doi: 10.1109/JSSC.2024.3505059.

[22] K. Cui, Y. Jin, X. Fan, and Y. Ma, "A 3-mV Precision Dual-Mode-Controlled Fast Charge Balancing for Implantable Biphasic Neural Stimulators," *IEEE Trans. Biomed. Circuits Syst.*, vol. 18, no. 4, pp. 896–907, 2024, doi: 10.1109/TBCAS.2024.3366518.

[23] Y. Oh, J. Hong, and J. Kim, "Integrated Low-Voltage Compliance and Wide-Dynamic Stimulator Design for Neural Implantable Devices," *Sensors 2023, Vol. 23, Page 492*, vol. 23, no. 1, p. 492, Jan. 2023, doi: 10.3390/S23010492.

[24] B. Garnier *et al.*, "A novel functional electrical stimulation sleeve based on textile-embedded dry electrodes," *Biomed. Eng. OnLine 2024 231*, vol. 23, no. 1, pp. 51-, Jun. 2024, doi: 10.1186/S12938-024-01246-8.

[25] S. Xu *et al.*, "Closed-Loop Wearable Device Network of Intrinsically-Controlled, Bilateral Coordinated Functional Electrical Stimulation for Stroke," *Adv. Sci. (Weinheim, Baden-Wurttemberg, Ger.*, vol. 11, no. 17, May 2024, doi: 10.1002/ADVS.202304763.

[26] J. W. Lee, E. Grabke, K. Chow, K. E. Musselman, and K. Masani, "Development of a closed-loop controller for functional electrical stimulation therapy plus visual feedback balance training for standing balance training," *Med. Eng. Phys.*, vol. 132, Oct. 2024, doi: 10.1016/J.MEDENGPHY.2024.104238.

[27] I. Brunner, C. B. Lundquist, A. R. Pedersen, E. G. Spaich, S. Dosen, and A. Savic, "Brain computer interface training with motor imagery and functional electrical stimulation for patients with severe upper limb paresis after stroke: a randomized controlled pilot trial," *J. Neuroeng. Rehabil.*, vol. 21, no. 1, Dec. 2024, doi: 10.1186/S12984-024-01304-1.

[28] M. A. Khan *et al.*, "A systematic review on functional electrical stimulation based rehabilitation systems for upper limb post-stroke recovery," *Front. Neurol.*, vol. 14, 2023, doi: 10.3389/FNEUR.2023.1272992.

[29] C. Höhler, E. Trigili, D. Astarita, J. Hermsdörfer, K. Jahn, and C. Krewer, "The efficacy of hybrid neuroprostheses in the rehabilitation of upper limb impairment after stroke, a narrative and systematic review with a meta-analysis," *Artif. Organs*, vol. 48, no. 3, pp. 232–253, Mar. 2024, doi: 10.1111/AOR.14618.

[30] L. R. L. Cardoso, V. Bochkezanian, A. Forner-Cordero, A. Melendez-Calderon, and A. P. L. Bo, "Soft robotics and functional electrical stimulation advances for restoring hand function in people with SCI: a narrative review, clinical guidelines and future directions," *J. Neuroeng. Rehabil.*, vol. 19, no. 1, Dec. 2022, doi: 10.1186/S12984-022-01043-1.

[31] C. SK and S. M, "The Effects of Functional Electrical Stimulation Cycling on Muscle Spasticity in Individuals With Spinal Cord Injury: A Systematic Review," *Top. Spinal Cord Inj. Rehabil.*, vol. 31, no. 1, pp. 77–99, Dec. 2025, doi: 10.46292/SCI23-00048.

[32] G. A. Hawker, S. Mian, T. Kendzerska, and M. French, "Measures of adult pain:

Visual Analog Scale for Pain (VAS Pain), Numeric Rating Scale for Pain (NRS Pain), McGill Pain Questionnaire (MPQ), Short-Form McGill Pain Questionnaire (SF-MPQ), Chronic Pain Grade Scale (CPGS), Short Form-36 Bodily Pain Scale (S…," *Arthritis Care Res. (Hoboken).*, vol. 63 Suppl 11, no. SUPPL. 11, Nov. 2011, doi: 10.1002/ACR.20543.

[33] N. Mohan, T. M. Undeland, and W. P. Robbins, "Power Electronics Convertors, Applictions, and Design," p. 824, 2002, Accessed: Nov. 08, 2026. [Online]. Available: https://www.wiley.com/en-us/power-electronics-converters-applications-and-design-3rd-edition-p-9780471226932

[34] B. M. Hasaneen and A. A. E. Mohammed, "Design and simulation of DC/DC boost converter," *2008 12th Int. Middle East Power Syst. Conf. MEPCON 2008*, pp. 335–340, 2008, doi: 10.1109/MEPCON.2008.4562340.

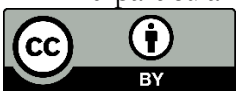
[35] D. W. Hart, "Circuit Simulation as an Aid in Teaching the Principles of Power Electronics," *IEEE Trans. Educ.*, vol. 36, no. 1, pp. 10–16, 1993, doi: 10.1109/13.204808.

[36] "Power Electronics - an overview | ScienceDirect Topics." Accessed: Nov. 08, 2026. [Online]. Available: https://www.sciencedirect.com/topics/engineering/power-electronics

[37] "(PDF) Current-Controlled Stimulator with Variable High Voltage Generation." Accessed: Nov. 08, 2026. [Online]. Available: https://www.researchgate.net/publication/336013151_Current-Controlled_Stimulator_with_Variable_High_Voltage_Generation

[38] "Gray's Anatomy: The Anatomical Basis of Clinical Practice." Accessed: Nov. 08, 2026. [Online]. Available: https://www.jinnahmedicalbooks.com/product/grays-anatomy-the-anatomical-basis-of-clinical-practice-42nd-edition/

[39] "Medical electrical equipment - Part 2-10: Particular requirements for the basic safety and essential performance of nerve and muscle stimulators | Standards Council of Canada." Accessed: Nov. 08, 2026. [Online]. Available: https://scc-ccn.ca/standards/notices-of-intent/csa-group/medical-electrical-equipment-part-2-10-particular-1



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