Design, Fabrication, and Experimental Validation of a Low-cost Neuromuscular Electrical Stimulator for the Prevention of Skeletal Muscle Atrophy

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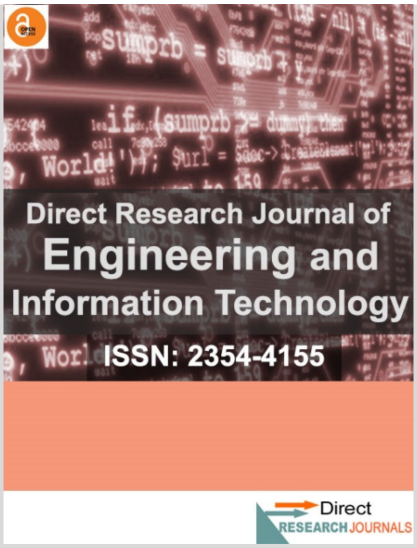
ABSTRACT

Individuals with limited mobility are at a significantly higher risk of muscle atrophy due to prolonged sedentary states. In rural Nigeria, the inconsistent electrical supply often renders contemporary electrical muscle stimulators (EMS) redundant. To address this, the current study aims to design and fabricate and experimentally validate a low cost, battery-powered muscle stimulator that ensures readily available therapy in resource-constrained environments. A user-centered design approach was adopted, involving the construction of a portable power management unit and a muscle stimulation circuit. The system's output parameters were verified using Proteus 8 simulation software to ensure voltage and current ranges aligned with established clinical standards. The hardware features an LCD interface that displays real-time frequency and stimulation duration. To evaluate clinical efficacy, handgrip strength gains were measured using a dynamometer and compared against a standard commercial EMS unit. The test was approved by the ethics committee of the Federal University of Science and Technology, Owerri. Nigeria. A paired-sample t-test conducted on 30 young adults between ages 18 – 30 revealed that the mean strength gain for the fabricated system (3.995 ± 0.483) did not differ significantly from the standardized EMS (3.947 ± 0.782), with t (29) = 0.516 and p = 0.610 (two-tailed). Furthermore, a strong positive correlation (r = 0.7899) indicated highly similar response patterns between the two modalities. These results demonstrate that the low-cost, battery-operated stimulator is a viable, high-performance alternative for muscle rehabilitation in areas lacking stable power infrastructure.

Keywords: *Muscle atrophy, muscle stimulator, battery-powered*



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INTRODUCTION

The therapeutic application of electrical stimulation (ES) has evolved into a cornerstone of modern rehabilitative medicine, making accessible a non-invasive means to initiate neuromuscular activity, reduce chronic pain, and improve functional outcomes across diverse pathologies (Allen et al., 2023). In acute care, neuromuscular electrical stimulation (NMES) has become a vital strategy for preventing muscle atrophy and accentuating physical recovery in intensive care units (Vila Pouca et al., 2025). Furthermore, the impact of bioelectrical pacing extends to the cellular level, where short-term stimulation has been shown to fundamentally improve the structural organization and maturation of heart tissues (Allen et al., 2024). However, despite these advancements, the clinical efficacy of ES is often restricted by the high cost of hardware and lack of specialized infrastructure in resource-limited environments.

The necessity for readily available NMES is most acutely felt in the management of neurological disorders, such as spinal cord injuries (Musselman et al., 2025) and post-stroke motor recovery (Khan et al., 2023). For patients navigating the long-term challenges of gait pathologies like foot drop, NMES provides essential orthotic and therapeutic support that improves daily mobility (Bleichner et al., 2026). Nevertheless, the "low-cost trap" in global health policy highlights a major barrier: households in developing regions often face health risks at different levels because affordable, high-quality medical devices remain inaccessible (Pei et al., 2025). To address this, healthcare resource management systems and models must now incorporate specific resource constraints into economic evaluations to ensure that medical interventions are both sustainable and scalable (Thokala et al., 2025; Eriskin et al., 2024).

Sustainable healthcare delivery in regions like Sub-Saharan Africa requires a significant shift toward digital health and decentralized care models. The integration of artificial intelligence (AI) and telemedicine has shown significant promise in bridging the diagnostic gap in the African health sector (Nnaji, 2024; Agbeyangi & Lukose 2025). However, these digital frameworks must be supported by robust, locally sourced and maintainable hardware. Current research suggests that intelligent machinery health management systems, typically used in industrial settings, offer transferable lessons for maintaining medical equipment in resource-constrained environments (Saeed et al., 2025).

From an engineering perspective, the frontier of neuroprosthetics is increasingly defined by "smart" wearable platforms. Recent innovations include electromyography (EMG)-driven closed-loop platforms for personalized pain modulation (Du et al., 2025) and functional electrical stimulation sleeves utilizing textile-embedded dry electrodes for enhanced user compliance (Garnier et al., 2024). While high-end systems such as soft neuroprosthetic hands that provide tactile feedback (Gu et al., 2021) or computer-controlled facial stimulators (Baker et al., 2024) demonstrate the technical potential of the field, their high cost necessitates the development of open-source options. Open-source designs for neuro-orthoses (Thorsen et al., 2025) and MRI-compatible sensors (Lubeck et al., 2024) have begun to democratize access to high-fidelity diagnostic and rehabilitative tools.

The sustainability of future medical devices also depends on novel application of resources and power solutions. Research into fully biodegradable, self-powered stimulators (Qu et al., 2025) and sustainable power modules for next-generation medical devices (Liang et al., 2025) reflects a growing emphasis on minimizing the environmental and economic burden of electronic waste. This is complemented by the development of customizable, low-cost nerve cuffs for selective neuromodulation (Riley et al., 2023) and modular simulators for specialized surgical training (Yeung et al., 2025). By making priority cost-utility and ease of fabrication (Nardi et al., 2025), engineers can build hardware designs that are as accurate as they are affordable (Liu et al., 2025).

To bridge the gap between clinical efficacy and economic viability, this study details the design, fabrication and experimental validation of a low-cost muscle stimulator. By building upon foundational principles of electronic muscle stimulator construction (Nzeribe et al., 2023) and integrating low-cost feedback systems (Freire et al., 2025), we aim to provide a versatile rehabilitative tool. Similar to low-cost designs developed for elbow rehabilitation (Jameel et al., 2024), the proposed device is engineered to deliver reliable muscle exercise while remaining accessible for local manufacture. The following sections outline the hardware architecture, the pulse-generation software, and the validation of the system as a sustainable solution for healthcare delivery in resource-constrained settings.

MATERIALS AND METHODS

Materials

The list of electronic components used are presented in (Table 1). Other components and devices used includes digital multi-meter, epoxy gum, cutter, soldering iron and lead was used as workshop tools.

Table 1: List of electronic components used.

S/N	Materials	Quantities	Specification	Source	Functions
1	Microcontroller	1	Arduino Nano	Local	Control task
2	Switching Transistor	1	MOSFET	Local	Pulse generator
3	Push Buttons	2	SPST	Local	Switching
4	Voltage Regulator	1	ICL7809	Local	Regulator
5	Switches	2	SPDT	Local	Switching
6	LCD Controller	1	I2C LCD	Local	Signal processing
7	LED	2	Blue & Red	Local	Indicator
8	Resistors	2	27K,1K,100Ohms	Local	Resistance
9	Diode	1	IN4007	Local	Rectification
10	Electrode	2	Surface	Local	Output
11	Potentiometer	1	10K	Local	Intensity Regulator
12	Plastic casing	1	1mm thick	Local	Casing
13	Rechargeable battery	1	lithium-ion	Local	Storage
14	Transformer	1	TRAN-2P2S	Local	Generate output

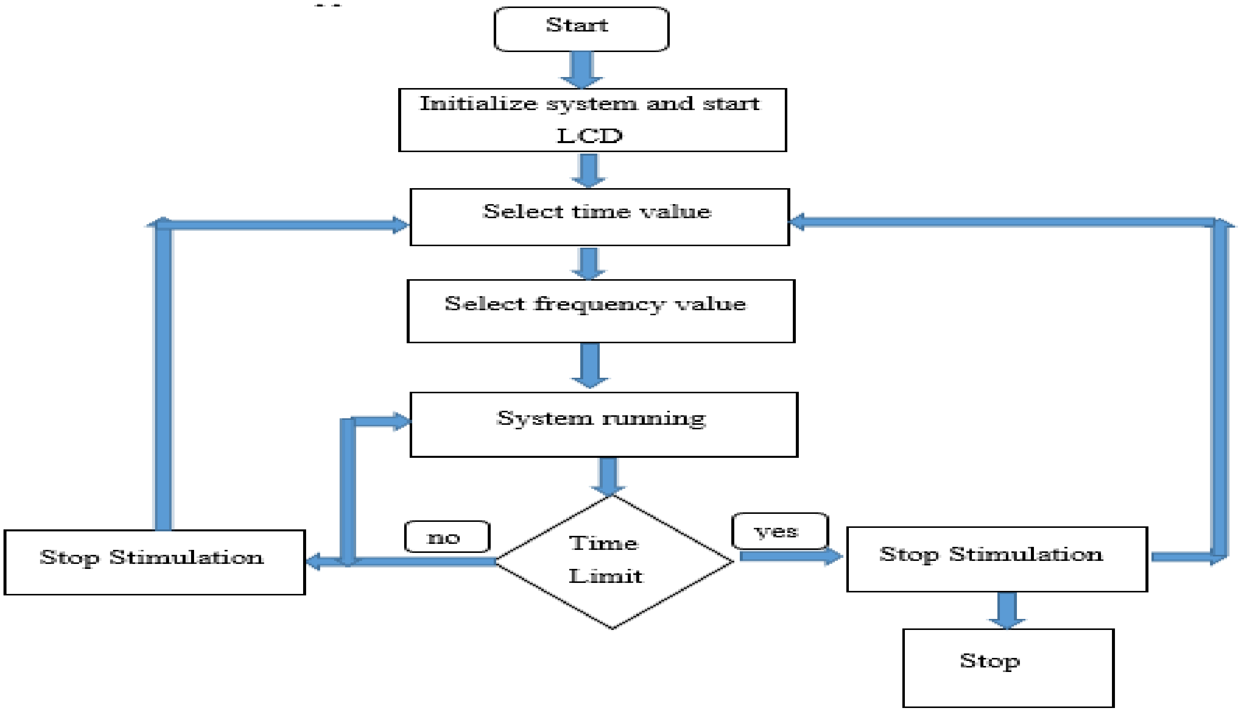

Figure 1: Algorithm Chart

Methods

Development on Bread and Vero Board

Methods employed in the design and construction a muscle stimulator begin by broadly classifying the entire system into two parts namely: Software and Hardware. The circuit was designed, test run and confirmed to be producing the appropriate output on a simulation software - Proteus 8. Arduino Integrated Development Environment (IDE) was used for the programming of the Arduino Microcontroller, while the hardware component was selected for been best fit for the purpose, connected and test run on a breadboard and transferred to a Veroboard.

Mode of Information Transfer and Algorithm Chart

The Algorithm for the programmed operation of the microcontroller is as follows:

1) Initialize the microcontroller
Clock Frequency = 18MHZ
Initialize I/O ports
Initialize LCD display

2) Option to input the frequency and duration value for the stimulation using the momentary push buttons.
3) The duration of stimulation ranging from 60seconds to 300seconds is determined.
4) The frequency of the stimulation from 5Hz to100Hz is determined.
5) Stimulation begins and stops after the time elapses.

The imputed microcontroller code for the Arduino Nano was written in C++ programming language. The Algorithm chart of the microcontroller program is contained in (Figure 1).

Components connectivity

The system is powered by a 12.1V Lithium -ion battery. A 7809 Linear Voltage Regulator steps down the battery voltage to a steady 9V DC. This powers the Arduino Nano and the I2C LCD driver. The "brain" of the system is an Arduino Nano microcontroller which manages the pulse-width modulation (PWM) and timing logic for the stimulation. A momentary Push Button is connected to A1 and A2 pins to allow user interaction. The I2C LCD (LM016L) provides real-time telemetry. The output stage is responsible for converting low-voltage DC into the high-voltage pulses required to elicit muscle contraction. The Arduino triggers the gate of an IRF640 N-Channel MOSFET (Q1) through a current-limiting resistor (R3 = 1kΩ) and a signal diode (D3). When the MOSFET switches off rapidly, the collapsing magnetic field in the transformer primary induces a high-voltage spike. Two LEDs (LED t and LED m) provide visual confirmation of the pulse frequency and power status. A step-up transformer is

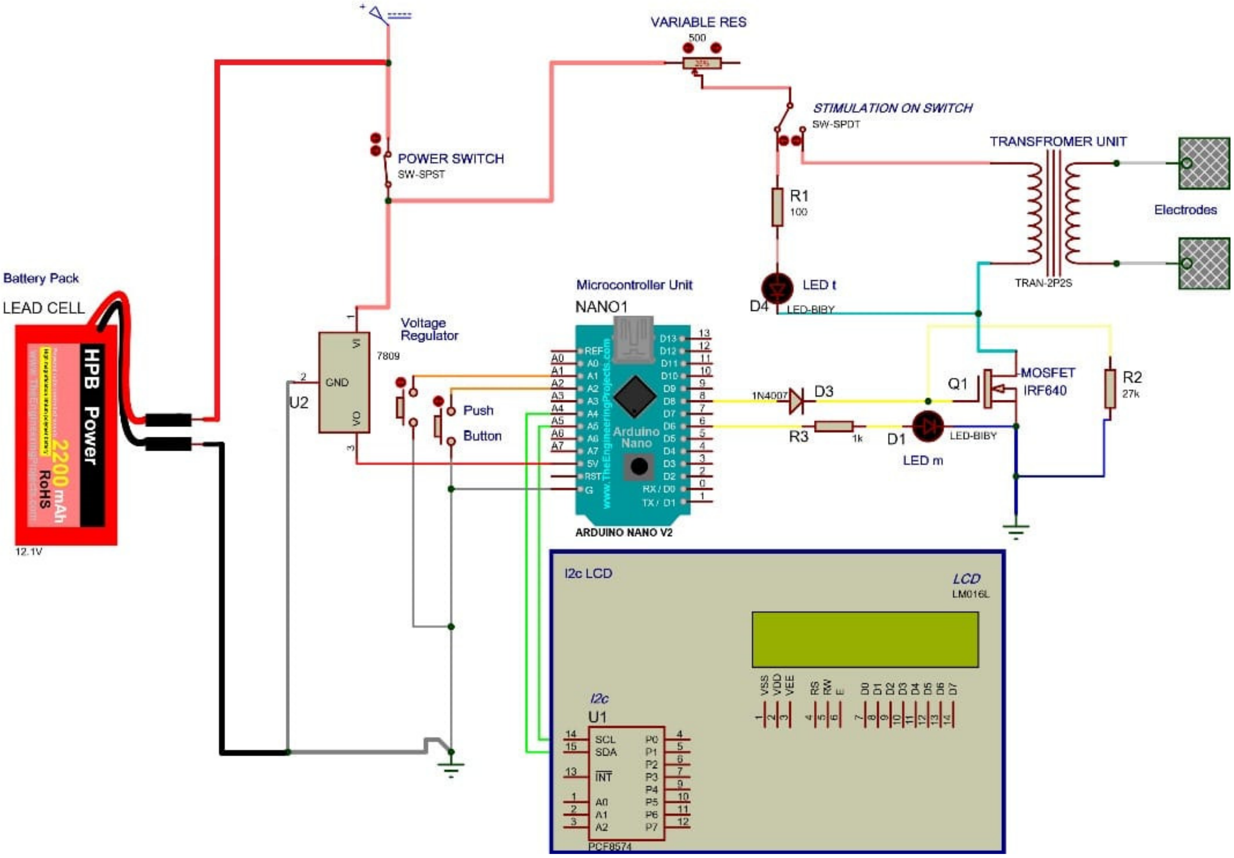

Figure 2: The Complete Circuit diagram

utilized to amplify the switched DC voltage. Having the primary side is connected to the regulated 9V rail and switched by the MOSFET, while the secondary side delivers the high-voltage biphasic or monophasic pulses to the electrodes. A 10KΩ Variable Resistor is placed in series with the transformer primary, allowing the user to manually attenuate the stimulation intensity by limiting the current flowing through the primary coil. The circuit includes critical safety features such as the SPST Power Switch which provides a disconnect for the entire control circuit. And the transformer which provides inherent electrical isolation between the control circuitry and the patient-contact electrodes (Figure 2).

The output voltage and current at the electrode were determined mathematically using Turns ratio (a) of the transformer and Ohms Law.

$$a = \frac{Vp}{Vs} = \frac{Is}{Ip} \tag{1}$$

$$V_S = I_S \times R \tag{2}$$

Where
a= 10
Vp= 12v
Vs=110v to 120v peak
Ip = 1 A peak (during ON pulse)
Is= 110mA peak
R= 10K ohm
Frequency range = 1Hz to 100Hz

This system makes use of a transformer to create amplified rectangular wave output, making the transformer our system's load.

Mechanical Structure

A plastic case was used for the fabrication of the Muscle Stimulator having a thickness of 1mm. Its 3D View is found in Figure 3 and dimensions in (Figure 4).

Data Collection

The inclusion criteria for all subjects for the study were that they must be young people aged 18 to 30 years and had intact, healthy skin at the electrode site. Participants were

Figure.3: View of a Muscle Stimulator

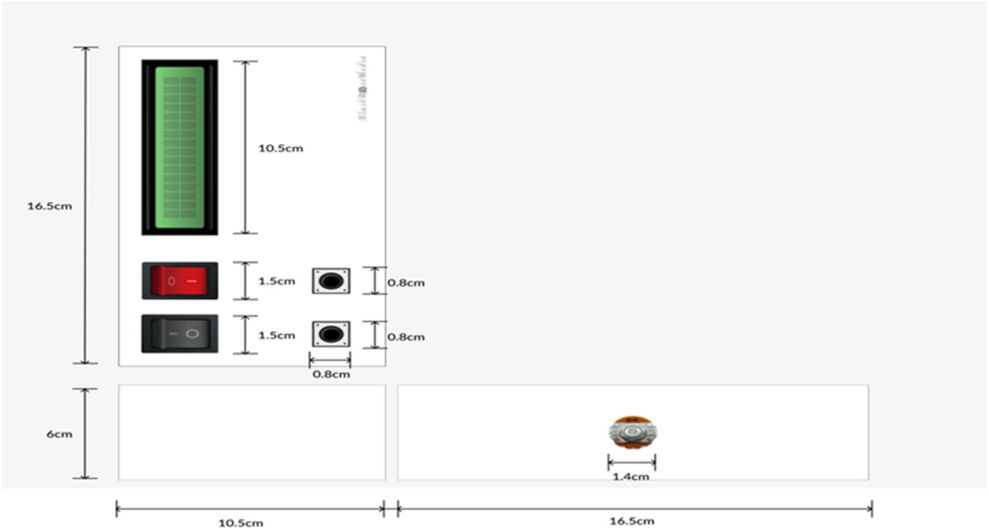

Figure 4: Dimensions on the muscle stimulator

excluded if they had in them a cardiac pacemaker, implanted cardioverter-defibrillator (ICD), or any other active electronic medical device. If they had a history of cardiac arrhythmias and recent myocardial infarction. Or have been diagnosed of peripheral neuropathy or loss of sensation at the target muscle site. Data was collected from the results of a handgrip strength test conducted on thirty (30) subjects at His Presence Physiotherapy Clinic, Port Harcourt. Nigeria. A hand grip strength test was conducted out on the subjects using a dynamometer (DIDIHOU, 200N, China) which was confirmed to be in good working condition before use.

Performance Test

The test was approved by the ethics committee of the Federal University of Science and Technology, Owerri. Nigeria. Each patient first performed a handgrip strength test before stimulation using the dynamometer and the value is noted. Then at the toss of a coin either the right or left hand was stimulated by the fabricated muscle stimulator while the other hand was stimulated using an already standardized muscle stimulator. The stimulation using both devices takes place at the same time for a duration of 5 minutes and a fixed frequency of 30Hz for all subjects. After which the subject would rest for 2 minutes before a handgrip strength test was performed. The strength gained on both hands, stimulated by the two devices was determined by subtracting the value of the strength before stimulation from the strength after stimulation.

RESULTS

Voltage and Current range of Stimulation

Mathematical derivation of Voltage output of system

This system makes use of a transformer to create amplified rectangular wave output, making the transformer our system's load. After calculating for the voltage and current output at the secondary end of the transformer using ohms law and the transformer ratio, the value obtained are tabulated in (Table 2).

Table 2: Voltage and Current range at the electrode.

Demonstration	Current (mA)	Voltage (volts)
Knob turned to the left (Min.)	1.19	5.69
Knob turned to the right (Max)	92.31	44.3

Output Signal using Oscilloscope

The output signals from the secondary end of transformer were observed at different frequencies on a digital oscilloscope on Proteous 8 software, displayed in (Figures 5 and 6). This output displays the amplified rectangular signal perceived as mild stimulation on the skin of the subject.

Overall System Performance

A hand grip strength test was conducted on 30 subjects at His Presence Physiotherapy Clinic in Port Harcourt, Nigeria as can be observed in Figures 7 and 8). A paired sample t-test was conducted to compare muscle strength gain under the fabricated Stimulator and the already standardized Electrical Muscle Stimulator (EMS). Results showed that the mean gain under EMS (3.995 ± 0.483) did not differ significantly from that of the fabricated device (3.947 ± 0.782), t (29) = 0.516, p = 0.610 (two-tailed). Although the fabricated device demonstrated a marginally higher average performance, the difference was not statistically significant. The strong positive correlation between both modalities (r = 0.7899) suggests similar response patterns among subjects. Figure 7 shows the model of correlation of the strength gained in the hands of the subject in the developed solar powered muscle stimulator against that of a standardized electrical muscle stimulator. The high magnitude of 0.7899 for R2 -value was obtained. This indicated high correlation of 78.9% and low variability of 21.1% in the measurement.

DISCUSSION

The present study demonstrated the successful design, fabrication, and validation of a low-cost, battery-powered electrical muscle stimulator tailored for use in resource-constrained environments. The selection of components listed in Table 1 reflects a deliberate emphasis on affordability, local availability, and functional adequacy. The integration of an Arduino Nano microcontroller, MOSFET switching transistor, I2C LCD controller, lithium-ion rechargeable battery, and adjustable potentiometer enabled programmable pulse generation, intensity modulation, and user feedback. This modular microcontroller-based architecture aligns with the rapid prototyping approaches reported by Luzio de Melo et al. (2015), who emphasized the flexibility and cost-effectiveness of microcontroller platforms in functional electrical stimulation (FES) systems. Similar to their framework, the present device leverages programmable control to regulate stimulation parameters while maintaining hardware simplicity.

The algorithm chart (Figure 1) and the complete circuit diagram (Figure 2) confirm that the system was logically structured to allow safe initialization, parameter selection, and controlled output delivery. The inclusion of voltage regulation (ICL7809) and rectification (IN4007 diode) ensured stable power conditioning, which is essential for preventing erratic stimulation. This design consideration is consistent with the broader bioelectronics principles discussed by Huang et al. (2024), who highlighted the importance of stable power management and precise waveform control in electrical stimulation devices. Although the present device is not implantable, its architecture reflects foundational bioelectronic system requirements, particularly in signal fidelity and user safety. The physical configuration of the muscle stimulator (Figures 3 and 4) demonstrates compactness and portability, features critical for deployment in rural or off-grid settings. Unlike implantable or micro-scale systems described by Cui et al. (2025), which focus on self-powered implantable platforms, the present study prioritizes external, rechargeable battery operation to address infrastructural challenges in developing regions. While implantable self-powered systems represent a cutting-edge direction in electrical stimulation, their complexity and cost may limit immediate applicability in low-resource settings. Therefore, the fabricated device fills an important translational gap between high-end biomedical innovation and accessible rehabilitation technology.

Electrical characterization results (Table 2) indicate that the output current ranged from 1.19 mA to 92.31 mA, while voltage ranged from 5.69 V to 44.3 V as the intensity knob was adjusted from minimum to maximum. These values fall within the typical therapeutic ranges reported for surface neuromuscular electrical stimulation, confirming that the fabricated device can deliver physiologically relevant stimulation. The wide adjustable range enables individualized therapy, an essential factor in neuromuscular rehabilitation. Ruiz-Gutiérrez et al. (2025) similarly emphasized the importance of tunable stimulation parameters in their custom electrical stimulation system for engineered muscle tissues. Although their application was in vitro and focused on 3D bioengineered muscle constructs, the principle of parameter optimization to elicit controlled muscular response is directly comparable to the present findings.

The oscilloscope outputs at 5 Hz and 30 Hz (Figures 5 and 6) further validate the functional integrity of the pulse generation circuit. Low-frequency stimulation (5 Hz) is

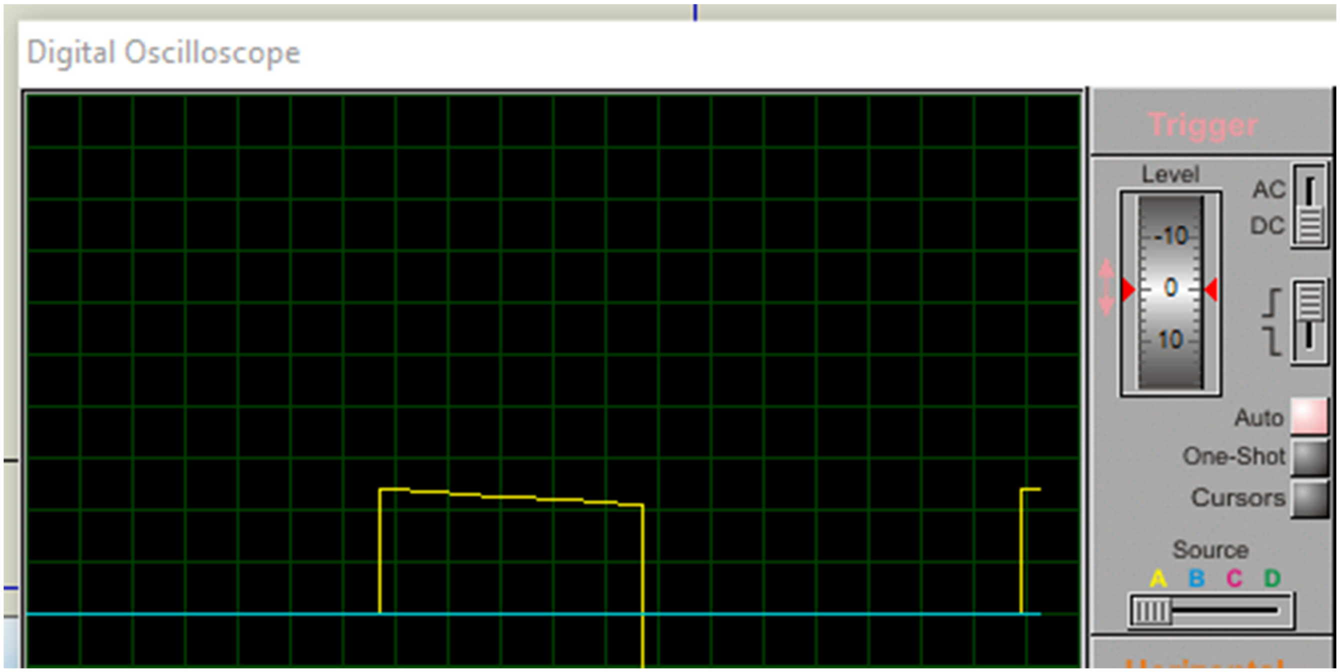

Figure 5: Output signal as show on the oscilloscope of Proteus 8 at 5Hz

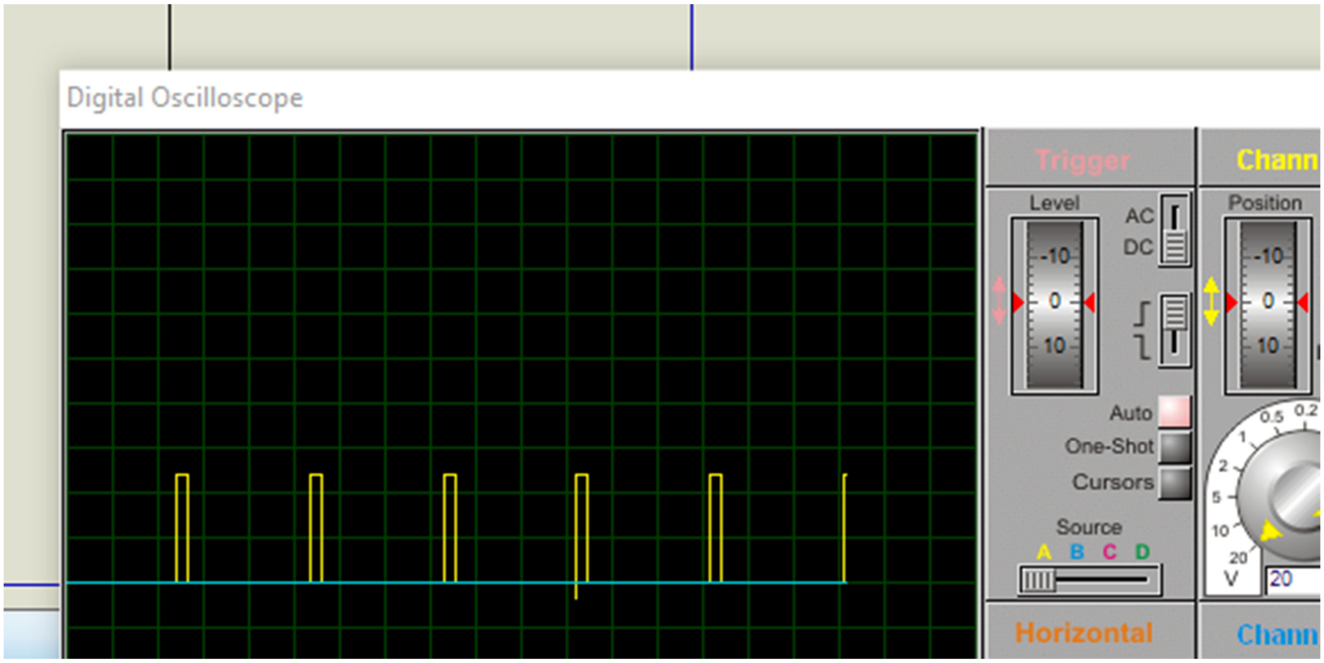

Figure 6: Output signal as show on the oscilloscope of Proteus 8 at 30Hz

typically associated with twitch contractions and endurance-type activation, whereas higher frequencies such as 30 Hz produce more sustained tetanic contractions. The clear waveform stability observed in Proteus 8 simulations confirms appropriate timing control by the microcontroller. This controlled frequency modulation parallels the precision requirements discussed by Angola Rojas (2020) in the development of implantable biohybrid muscle stimulation systems, where accurate temporal control is critical for effective neuromuscular activation. While Angola Rojas focused on implantable applications for lower motor neuron injury, both systems underscore the necessity of reliable frequency regulation to ensure predictable muscle response.

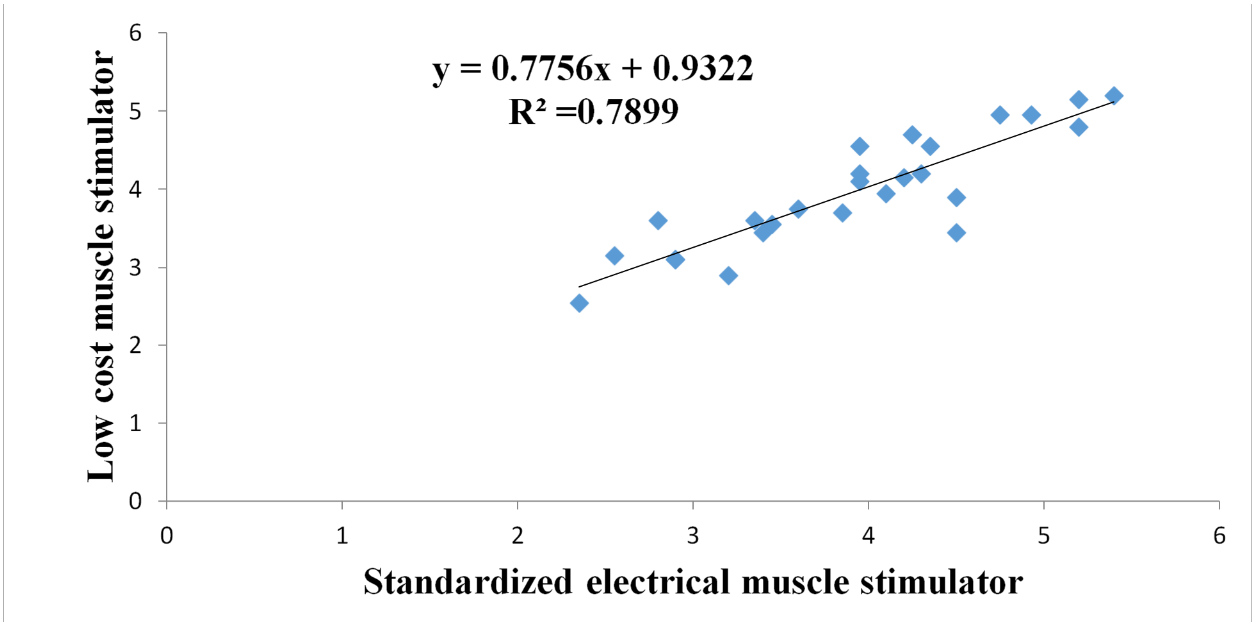

Figure 7: Correlation Model of muscle strength gain in the Low-cost vs the Standard Electrical Muscle Stimulator.

The statistical evaluation of muscle strength gain provides further empirical support for the device's clinical relevance. The paired sample t-test showed no significant difference between the fabricated device (3.995 ± 0.483) and the standard EMS (3.947 ± 0.782), t(29) = 0.516, p = 0.610. This statistical equivalence suggests that the low-cost stimulator achieves comparable functional outcomes to commercially available systems. The strong positive correlation (r = 0.7899) illustrated in Figure 7 reinforces this finding, indicating consistent physiological response patterns across both modalities. From a translational research perspective, this comparability is highly significant, as it demonstrates that cost reduction does not necessarily compromise therapeutic efficacy.

These findings resonate with the validation-oriented approaches of Parkhill (2024), who emphasized rigorous functional testing in the design of a 3D-printed muscle stimulation chamber. Although Parkhill's work addressed in vitro tissue culture environments, both studies share a methodological emphasis on experimental validation rather than purely theoretical design. Similarly, Camps Carré (2022) highlighted the importance of accurate force measurement systems in evaluating intramuscular microstimulators. The use of a dynamometer in the present study (Figure 9) aligns with such biomechanical validation practices, ensuring that muscle strength gain was objectively quantified rather than subjectively assessed.

The real-time application of the device during muscle stimulation (Figure 8) confirms its operational feasibility in a practical setting. In contrast to advanced electroactive biomaterial systems described by Yang et al. (2025), which integrate smart materials for soft tissue engineering, the present device relies on conventional surface electrodes and discrete electronic components. However, both approaches share a common objective: enhancing neuromuscular activation through controlled electrical cues. The key distinction lies in technological complexity and target context; whereas Yang et al. (2025) focus on regenerative engineering applications, this study addresses accessible rehabilitation for immobile individuals in underserved communities.

The integration of user-adjustable intensity control and LCD feedback also reflects a human-centered design philosophy comparable to assistive device innovations such as Abdullah (2023) biofeedback technology development. Although Abdullah's work targeted mobility aids, both studies underscore the importance of usability, feedback, and patient autonomy in biomedical device design. By allowing users to monitor frequency and duration, the present stimulator enhances therapeutic engagement and operational transparency.

Overall, the study contributes to the growing body of literature on electrical stimulation technologies by demonstrating that functional equivalence to standard EMS systems can be achieved through locally sourced, low-cost components. While high-end implantable and biohybrid systems (Cui et al., 2025; Angola Rojas, 2020) represent the frontier of biomedical innovation, scalable and affordable external stimulators remain crucial for global health equity. The comparable muscle strength gains observed in this study substantiate the argument

Figure 8: Muscle Stimulation Ongoing.

Figure 9: Dynamometer in use

that technological sophistication must be balanced with contextual appropriateness. In this regard, the fabricated device represents a pragmatic and sustainable advancement in neuromuscular rehabilitation, particularly for regions with limited infrastructure and financial resources.

Conclusion

Individuals in immobilized states are in dire need of muscle stimulation so as to prevent muscle atrophy that may need to other medical complications. The developed low-cost muscle stimulator has proven to be a suitable for this objective especially for rural areas without sustainable electrical power supply.

Recommendations

The sample size (n = 30) restricts the generalizability of the findings. Additionally, the duration of intervention was relatively short, and long-term effects such as endurance, fatigue resistance, and user comfort were not examined. Future studies should incorporate larger cohorts, longitudinal assessments, and optimization of energy storage and control mechanisms to improve reliability. Integration of IoT-based monitoring could further advance the device into a smart rehabilitation device for remote healthcare applications.

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