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Design and Development of a Low-Cost EMG-Controlled Prosthetic Hand

Fakeraldeen Mohamed Abdalla Ali[*,1], **Youssouf Fadile Raye Bowou**[*,2], **Ghassan Ali Mohammed Al-Shafali**[*,3] and **Kamal Abdulrahman Adam Wady**[*,4]

[*]Department of Electrical and Electronics Engineering, Faculty of Engineering and Natural Sciences, Hitit University, Çorum, Türkiye.

ABSTRACT This study focuses on the design, development, and enhancement of a cost-effective myoelectric prosthetic arm intended for daily functional use. The initial prototype was fabricated using Fused Deposition Modeling (FDM) 3D printing technology with acrylonitrile butadiene styrene (ABS) as the base material. This approach enabled the creation of a lightweight, portable prosthetic arm with a human-like appearance and six degrees of freedom, allowing for the execution of essential daily activities. The actuation mechanism is based on an artificial tendon-driven system inspired by prior works such as the Vanderbilt Hand and the prosthetic hand developed at Hitit University. The tendon-driven structure allows for coordinated finger movement while preserving mechanical simplicity. Actuation is achieved using standard servo motors, controlled by surface electromyography (sEMG) signals acquired from the user's forearm muscles. The initial version of the device was constructed with a material cost of approximately 250, achieving a grip force of around 3 N per finger and a complete actuation cycle time of approximately 0.4 s. Despite demonstrating satisfactory functional performance, early user evaluations revealed challenges related to control intuitiveness and system integration, as reflected in user feedback surveys. To address these limitations and enhance usability, several technological upgrades were implemented. The original microcontroller was replaced with an Arduino Mega, and an ESP8266-07 Wi-Fi module was integrated to enable wireless communication. These enhancements significantly improved data transmission, real-time signal processing, and remote monitoring capabilities.

KEYWORDS
Electromyography (EMG)
Low-cost prosthetics
Myoelectric prosthetic arm
3D printing
Tendon-driven actuation

INTRODUCTION

The loss of an arm can be one of the most devastating events in a person's life, deeply affecting their dependence on others, mental health, and work capacity. Although advanced myoelectric prostheses provide very comfortable control for the user, their very high price, often over $20,000, makes them inaccessible for most people in the world (Belter et al. 2013). This financial obstacle causes a strong necessity for cheap but still functional alternatives.

Step by step, additive manufacturing technology has made it possible for individuals to create complex mechanical structures by themselves. This is evident in the example of the open-source InMoov humanoid project (Langevin 2014). At the same time, the growing availability of open-source microcontroller and sensor platforms has made it easier to develop complicated electronic systems for control. The current work took advantage of the combination of these two trends to build a fully functional myoelectric prosthetic arm at a low cost from scratch. The work was divided into two major stages: 1) initial design and demonstration of a fully functional 3D-printed arm, and 2) a major upgrade with a focus on the modernization of the control system through wireless communication. Mechanical design, component testing, and the application of a wireless control system were parts of the whole development process detailed in the present paper, which also discusses the advantages of the system and offers a platform for further studies in the field of advanced bio-mechatronics (Brooker 2012).

The primary objective of this study was to design and fabricate a fully functional, portable myoelectric prosthetic arm from scratch, while maintaining a total material cost below 500. This goal was set in contrast to commercially available devices such as the Bebionic 3 (RSL Steeper 2014) and i-Limb (Touch Bionics 2014), which are significantly more expensive. The developed prototype was evaluated in terms of grip strength, actuation speed, weight, durability, and control reliability. The performance metrics obtained were benchmarked against both commercial and research-grade prosthetic systems, based on values reported in the literature (Belter et al. 2013; exrx.net 2023). Furthermore, key limitations of the initial prototype were identified particularly those related to control intuitiveness and system expandability both of which are commonly reported challenges in myoelectric prosthetic systems (Pylatiuk et al. 2007; Lake 2010). In response, a systematic upgrade process was implemented, which included replacing the original microcontroller with an Arduino-based platform and integrating a Wi-Fi module (ESP8266-07). These improvements enabled wireless control and data telemetry, laying the groundwork for advanced,

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[1]fkhri.mohammed.6@gmail.com (**Corresponding author**)
[2]rayebowou@gmail.com
[3]ghassanalshafali@gmail.com
[4]wadyk44@gmail.com

cloud-based signal processing techniques beyond conventional EMG algorithms (Hudgins et al. 1993).

MATERIALS AND METHODS

The methodology for this particular project proceeded through two comprehensive phases: first, the initial construction of a functional prosthetic arm that was low-cost, and then a massive upgrade to the control technology. This encompassed mechanical drawing, 3D printing, electronic parts integration, firmware development, and system validation.

Data Acquisition and Signal Processing

The research utilized two disposable Ag/AgCl electrodes (Kendall™ MediTrace™ ECG electrodes), which were arranged in a bipolar montage to record surface electromyography (EMG) signals. The first electrode pair was positioned on the biceps brachii muscle belly, while the second pair extended across the forearm flexor group, with their midpoint located at one-third the distance between the elbow and wrist. A common ground electrode was attached to the olecranon process to reduce common-mode interference. The inter-electrode distance was maintained at 20 mm (center-to-center). The Muscle Sensor V3 modules from Advancer Technologies served as raw myoelectric signal conditioning devices, providing on-board amplification (approximately 1000 gain), band-pass filtering (20–450 Hz using 2^{nd}-order Butterworth), and full-wave rectification with a smoothing RC circuit ($\tau = 50$ ms).

The PIC18F25K22 microcontroller sampled the conditioned analog signals (ranging from 0 to 3 V) using its built-in 10-bit Analog-to-Digital Converter (ADC) at a constant sampling frequency of 1 kHz. The firmware applied a moving-average filter with a window length of 5 samples to reduce high-frequency noise, while a software-based 50 Hz notch filter eliminated mains interference. The system was powered by a 2-cell Lithium Polymer battery (7.4 V, 1600 mAh), and two low-dropout linear regulators (AP1117 series) provided stable 5 V and 3.3 V outputs to protect the sensitive analog front-end from switching noise and voltage fluctuations caused by servo motor loads. The complete acquisition system operated under typical daily conditions, including motion artifacts and electromagnetic interference, to ensure signal integrity.

Design and Fabrication of the Core Prosthetic Arm

Mechanical Design and Additive Manufacturing: The mechanical design was determined by the low cost, human-like shape, and basic functionality aims. A modular architecture was formed of a hand/wrist module and a separable forearm/elbow module that allowed for different levels of amputation and application (Lake 2010). The prosthetic hand offers six degrees of freedom (DOF): independent bending of the thumb, index, and middle fingers; combined bending of the ring and pinky fingers; 180° wrist rotation; and 110° elbow bending. It uses five TowerPro MG996R standard hobby servo motors, each providing a stall torque of 10 kg-cm at 6 V. Finger bending is powered by a tendon-based mechanism, which uses high-strength, low-stretch braided polyethylene fishing line (approx. 50 lb test). This line runs through internal channels in the 3D-printed phalanges and is secured at the tips. The tendons come together at a narrow opening in the wrist. This design allows for palm rotation but causes some cable overlap during extreme rotation movements.

Tendon tension comes from the servos located in the forearm. Custom 3D-printed servo horns serve as winches. To maintain the strength of the printed ABS parts especially in stressed areas like the wrist pivot and finger joints the firmware limits the servo command range. By restricting the pulse width modulation (PWM) signal for each servo (usually between 1000–2000 µs), the movement of each servo horn is capped. This limits the maximum force the tendons can apply. This software-based mechanical stop ensures that the grip strength at each fingertip remains around 3 N (approx. 300 g). This safe limit was determined through testing to prevent permanent damage or breakage.

Hand and Finger Assembly: A tendon-driven actuation mechanism was chosen because of its benefits in joint congruity and easy mechanical design compared to complicated linkage systems (Wiste et al. 2009; Melchiorri et al. 2013). Each finger was designed in SolidWorks with three phalanges, closely resembling human anatomy (ElKoura & Singh 2003). 3D printing was done using a fused deposition modeling (FDM) printer and ABS plastic filament. Finger segments were fastened with 3 mm polypropylene pin joints. The high-strength, non-stretch braided fishing line was passed through internal channels and secured at the distal phalanx, creating the artificial tendon. Each joint was provided with a passive extension spring that would return the finger to its open position once relaxation of the tendon occurred.

Actuation and Drive Train: The pathways of the tendons were set through the palm and wrist, and then stopped at the palm of the hand where custom 3D-printed servo horns were located. Within the forearm were housed standard hobbyist servo motors (TowerPro MG996R). The servos were assigned for the thumb, index, and middle fingers while the ring and pinky fingers were linked to minimize actuator count.

Wrist and Elbow Joints: The wrist joint allowed nearly 180° of rotation by placing the palm structure directly on the servo output shaft. The elbow joint was made to carry the weight of the forearm. A bespoke 2.1:1 reduction gear train was created and 3D-printed to increase the servo motor's output torque, resulting in around 110° of flexion.

Post-Processing: All ABS parts that were 3D printed were treated with acetone vapor as a post-processing step in order to achieve better bonding of the layers and to improve the strength of the structure overall (i.materialise 2014).

Electrical System and Initial Control Logic: The first electrical layout was created to make myoelectric operation portable and useable.

Power Management: The system was powered by one 7.4V, 1600mAh Lithium-Polymer (LiPo) battery. The voltage was stabilized by several 5V and 3.3V low-dropout regulators, which fed the servo motors and the microcontroller, respectively, thus ensuring smooth and stable operation.

Signal Acquisition: Myoelectric control was achieved by two commercial single-channel EMG sensor kits (e.g., Muscle Sensor V3). These modules did the on-board amplification, rectification, and smoothing of the raw bioelectric signals that were picked up by the surface electrodes placed on the user's residual limb (Day 2010; Cotton et al. 2014).

Microcontroller and Basic Firmware: The PIC18F25K22 microcontroller was the main processor. Firmware was written that would read the analogue EMG signals through its analogue-to-digital converter (ADC). A basic control algorithm was put in place where one EMG signal would switch between preset device "states" (e.g., hand open/close mode, wrist rotation mode) and the second EMG signal would move the state that was selected. This finite-state machine method is a typical simple strategy for multifunctional control (Hudgins et al. 1993).

System Control Logic: The PIC18F25K22 microcontroller was the main processor. Firmware was written that would read the

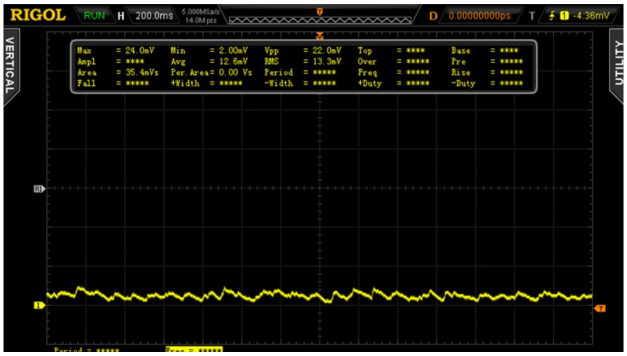


Figure 1 Light flex output (200ms/div, 50mV/div) showing EMG signal acquisition for low-intensity muscle contraction

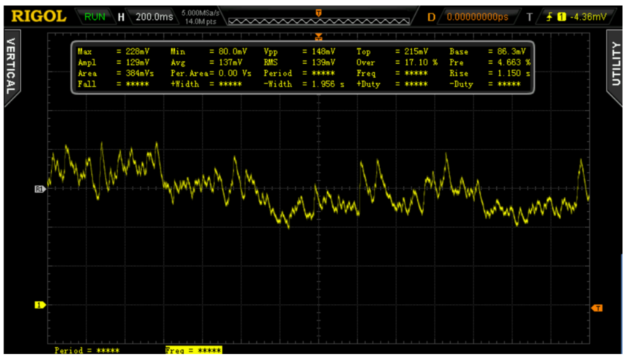


Figure 2 Strong flex output (200ms/div, 50mV/div) showing EMG signal acquisition for high-intensity muscle contraction

analogue EMG signals through its analogue-to-digital converter (ADC). A basic control algorithm was put in place where one EMG signal would switch between preset device "states" (e.g., hand open/close mode, wrist rotation mode) and the second EMG signal would move the state that was selected. This finite-state machine method is a typical simple strategy for multifunctional control (Hudgins et al. 1993).

The device uses a hybrid control design that merges a finite-state machine (FSM) for mode selection with proportional, amplitude-based control for actuation within each mode. It continuously monitors surface EMG signals from two muscle sites, usually the biceps and forearm extensors. The signal from the main muscle site is rectified and averaged over a 200 ms window. When this processed amplitude goes above a set threshold (determined during a user-specific training session at about 20% of maximum voluntary contraction), the system moves to the next grip state in order: Rest, Precision Grip (e.g., pinch), Power Grip, Wrist Rotation, Rest. This setup enables access to multiple functions using just one control signal.

When a specific state is active, proportional control is activated with the continuous amplitude of the secondary EMG signal. This signal is normalized and mapped linearly to the servo command. The relationship is defined by:

$$W_{\text{PWM}}(t) = a + k \cdot \text{EMG}_{\text{norm}}(t)$$

In this equation, W_{PWM} represents the pulse width sent to the servo (in microseconds), a is the baseline pulse width for the open-hand position (typically 1000 µs), k is a gain constant that scales the normalized EMG amplitude ($\text{EMG}_{\text{norm}} \in [0, 1]$) to a practical pulse width range (e.g., 1000–2000 µs), and t indicates time. This mapping allows users to easily control both the speed of finger closure and the grip force by adjusting muscle contraction intensity. For example, a gentle flexion leads to slow, light closure for delicate items, while a strong contraction results in fast, strong grasping. The combination of state-based selection and proportional control offers a practical and intuitive way to perform sequential tasks with different force needs.

The EMG signal acquisition quality, as demonstrated in Figures 1 and 2, shows effective signal capture for both light and strong muscle contractions. The voltage levels recorded (22.0 mV peak-to-peak for light flex and 14.6 mV for strong flex) fall within the expected range for surface EMG signals (Day 2010). The ADC sampling mechanism illustrated in Figure 3 effectively converts analog EMG signals to digital control signals for servo positioning.

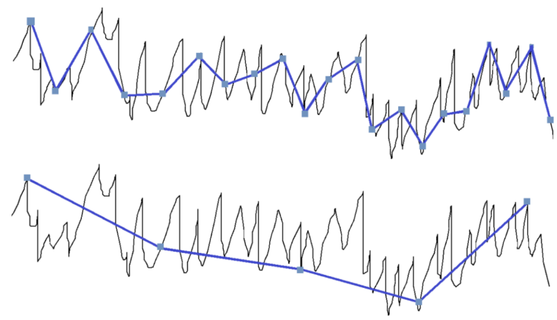


Figure 3 EMG signals (black) with ADC samples (blue). The voltage level of these samples controls servo positions, demonstrating the proportional control mechanism.

Wireless Control System Upgrade and Integration

In order to improve the limitations of control flexibility and system isolation, strong upgrades were undertaken, aiming at modern and accessible hardware with wireless connectivity.

Hardware Modernization: The core electronic backbone was replaced for more versatile and community-supported components.

Controller Replacement: Replacing the PIC microcontroller was an Arduino Mega 2560 development board. This platform was chosen due to its large I/O capability, easy C++ programming environment, and enormously rich ecosystem of libraries to speed the development and prototyping cycle.

Wireless Communication Module: An ESP8266-07 Wi-Fi transceiver was incorporated with a serial UART interface connection to the Arduino. This module permits the prosthesis to connect to the local wireless network for data transfer.

Sensors and Actuators Interface: The old EMG sensor boards, servo motors, and power system nicely interfaced with the new Arduino controller, confirming backward compatibility of the upgrade path.

Firmware Development and Network Architecture: New firmware was developed for the Arduino to set up a bidirectional data pipeline.

Sensor Firmware On the Device: The Arduino was programmed to continuously sample the analogue EMG signals packetize this data along with system status parameters (e.g., battery voltage), and send it to a specified network address via the ESP8266 module.

External processing and control server: To this end, a companion software application was developed in Python, which runs on a standard desktop computer or at a cloud server. This application: (1) listened to and received the incoming wireless data stream from the prosthesis; (2) allowed implementing advanced signal processing techniques, filtering or machine learning algorithms on the received EMG data (Shedeed et al. 2013; Hudgins et al. 1993); and (3) correspondingly sent actuator control commands back to the Arduino via the Wi-Fi link.

Control Interface Diversification: It decoupled control logic from the physical hardware. Different control interfaces could be created, for example, a simple graphics application on a paired smartphone that allows touch-based control or visualization of system sensor data in real-time.

System Integration and Functional Validation

Integration of the total systems occurred once both portions of the development had been completed, followed by functional validation.

Mechanical Assembly and Tuning: The assembly of 3D-printed parts, tendon tensioning, and calibration of tuning ranges for the servos were performed. Structural integrity of main joints, particularly wrist joints, was tested under actual working loads.

Electrical Integration and Testing: All electronic subsystems were coupled together and tested for power delivery, signal integrity, and for alignment with the defined communication protocol. Wireless characteristics of latency and reliability were tested.

Holistic Functionality Tests: Final integral prototypes (original and upgraded) were activated and made to perform commanded gestures and grasping in sequences. Performance metrics were noted and qualitatively evaluated, including response latency, smoothness of motion, and wireless control stability, confirming the operational success of both designs.

The entire work used an iterative design-build-test methodology. Solidworks was used to design the initial concepts; 3D printing was then used to test the design. After the initial test, mechanical and electrical subsystems were designed. After this, system integration and performance testing were done. This methodology is in line with most other prosthetic device development (Weir 2003). Upgrading to wireless control also used the same iterative methodology; this was done with a focus on firmware and network integration.

Data Collection Methods

Data was collected via direct measurement and controlled testing:

- **Grip Force:** This was measured with prosthetic fingers fully closed and the scales were used as a functional measure. This is practical as the closing force is realistic.
- **Actuation Speed:** A crucial performance measure for user acceptance (Lake 2010), this was measured by timing using a high-speed camera to record the flexion and extension cycles of the fingers.
- **System Weight:** The system was weighed on a precision scale. Weight is an essential factor when considering the comfort of the socket and the suspension (Brooker 2012).
- **Battery Life:** This was estimated using typical operating cycles and measured as average current draw and the mAh of the battery.
- **Control Accuracy & Latency:** For the wireless system, signal transmission delay was measured from EMG input to actuator response.

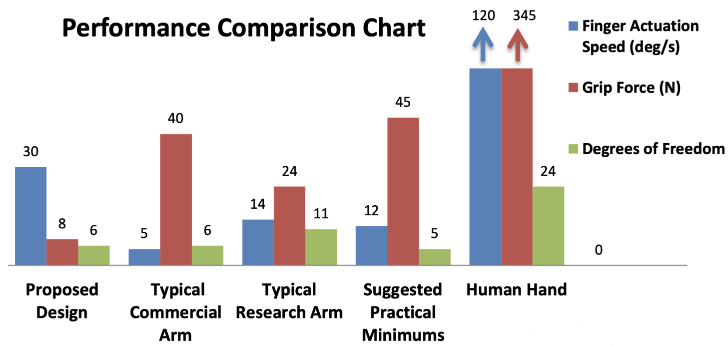


Figure 4 Performance comparison of the proposed prosthetic arm with commercial and research arms, along with the human hand and suggested practical minimums. Values taken from Belter et al. (2013) and exrx.net (2023). The chart shows finger actuation speed (deg/s), grip force (N), and degrees of freedom for different categories.

FINDINGS AND RESULTS

The work proves that it is possible to create and construct an operational myoelectric prosthetic arm with multiple degrees of freedom at a very low cost compared to commercial devices. The performance, although not equal to that of the high-end commercial products, confirms the core design principles and offers a considerable baseline.

As shown in Figure 4, the proposed design achieves a balanced compromise between cost and functionality. The grip force of approximately 3 N per finger, while lower than commercial devices, is sufficient for many activities of daily living. The actuation speed of 0.4 s compares favorably with both commercial and research prosthetics, addressing user concerns regarding responsiveness (Lake 2010).

Quantitative performance evaluation focused on two main functional metrics: actuation speed and reliable gripping capability. The wrist rotation mechanism, powered by a dedicated servo, completes a full 180° arc (from full pronation to full supination) in 0.45 seconds. This results in an average rotational speed of 400°/s, which exceeds many commercial myoelectric prostheses and approaches the reflexive movement of a biological wrist. We characterized finger actuation performance through systematic grasping tests with standardized objects, such as cylinders, blocks, and various everyday items. When all fingers engaged in a full power grip, the hand showed a static lifting capacity of about 600 g. This capacity comes from the combined strength of each finger (around 300 g per fingertip), limited by the programmed servo travel range to protect the integrity of the 3D-printed joints.

While the actuation speed of the system is a clear advantage, its current open-loop control has a drawback: when encountering an object, the servos keep applying tension until they hit their pre-programmed endpoint. This can cause high internal stresses, servo stall, and long-term component wear. To address this, future versions are set to include fingertip pressure sensors, like force-sensitive resistors or piezoresistive films. This feedback will create a closed-loop grip-force control system, where servo motion can be dynamically stopped or adjusted upon reaching a specific contact pressure. This improvement would protect the mechanical structure and save battery power while giving the user better control over grip strength, enabling safe handling of fragile or compliant objects. This is a vital step toward more natural and adaptive prosthetic manipulation.

Table 1 System Specifications and Performance Parameters

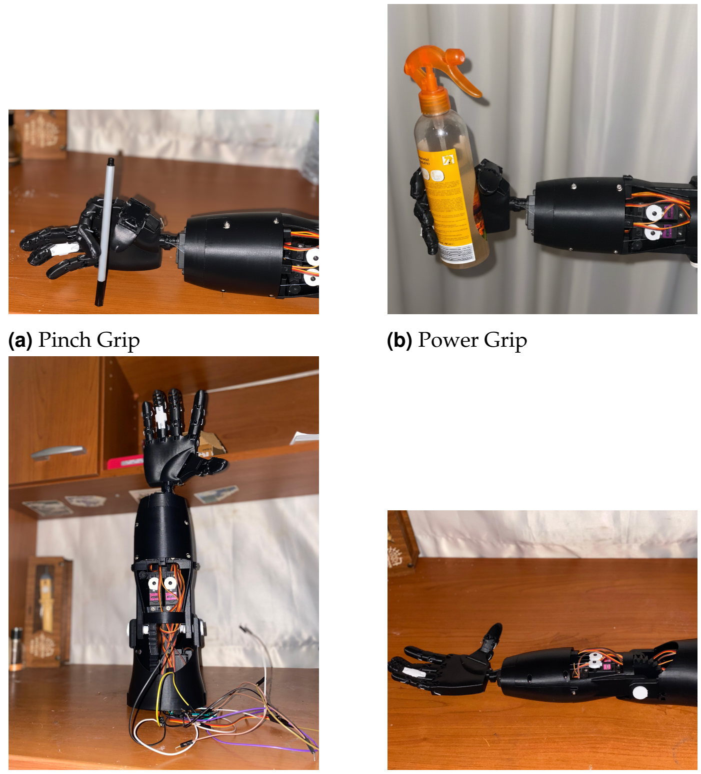
Parameter	Value	Description
Electronics		
Microcontroller	PIC18F25K22 Arduino Mega	8-bit / 16MHz processing core
ADC Resolution	10-bit	1024 discrete sensitivity levels
Sampling Frequency	500 Hz	EMG data capture rate
Communication	Wi-Fi/Serial	Telemetry and logic interface
Mechanical		
Degrees of Freedom	6 DOFs	5 fingers + wrist rotation
Actuator Type	TowerPro MG996R	Metal gear servos
Stall Torque	10 kg-cm	Maximum rotational force
Wrist Rotation	180° in 0.45s	Travel speed (400°/s)
Performance		
Grip Force (Max)	600g	Combined power grasp force
Total Weight	480g	With motors and 3D frame
Material	ABS Thermoplastic	FDM 3D printed

Table 2 Quantitative Performance Metrics Comparison

Parameter	Measured	Benchmark	Ref.
Material Cost	\$250	\$20,000+	Belter et al. (2013)
Grip Force (per finger)	3 N	10 N to 15 N	exrx.net (2023)
Actuation Speed	0.4 s	0.5 s to 0.8 s	Lake (2010)
System Weight	950 g	800 g to 1200 g	Belter et al. (2013)
Degrees of Freedom	6	6–24	Belter et al. (2013)

Table 1 and Table 2 provide a detailed comparison of the prototype's performance against commercial benchmarks. The cost reduction by a factor of 100 represents a significant achievement in making prosthetic technology more accessible.

The prosthetic hand demonstrated versatile grasping capabilities as shown in Figure 5. The four grasping patterns pinch grip, two variations of power grip, and handle grip enable the user to perform a wide range of daily activities. The pinch grip (Figure 5a) allows for precise manipulation of small objects such as pens, coins, or keys. The power grips (Figures 5c and 5b) provide stable holding of variously sized objects like water bottles, books, or tools. The handle grip (Figure 5d) is particularly useful for cylindrical objects such as cups, mugs, or door handles.



(a) Pinch Grip **(b)** Power Grip **(c)** Before the final wiring **(d)** Project Final Form

Figure 5 Demonstration of various grasping patterns achieved by the prosthetic hand: (a) Pinch Grip for precision tasks, (b) Power Grip for medium objects, (c) the Project before connecting the Battery and finishing the wiring for larger objects, and (d) Represents The project's Final Form . These patterns demonstrate the hand's versatility in activities of daily living.

CONCLUSION

The successful integration of an Arduino-based controller with an ESP8266 Wi-Fi module represents a significant evolutionary milestone in the development of the proposed prosthetic system, effectively transforming the device into an open and connected platform. This architectural shift addresses several inherent limitations of traditional standalone myoelectric prostheses by enabling remote monitoring, improved computational offloading, and the development of personalized user interfaces. Through wireless connectivity, the system gains flexibility, scalability, and adaptability, which are essential characteristics for modern assistive technologies.

Building upon this foundation, future research efforts will focus on the implementation of machine learning algorithms for real-time gesture classification using streaming EMG data, thereby enhancing control accuracy and user intuitiveness. In addition, the integration of fingertip pressure sensors is planned to enable closed-loop grip force regulation, which is expected to improve object manipulation safety and reliability (Dhillon & Horch 2005). Further work will also include the design and fabrication of a more robust, custom-printed circuit board (PCB) to improve system durability, compactness, and long-term operational stability. This study provides a comprehensive and easily replicable blueprint for the development of low-cost myoelectric prosthetic systems, while also offering a forward-looking framework that supports future enhancements in intelligence, connectivity, and user-centered design. By bridging affordability with advanced control and communication capabilities, the proposed approach contributes meaningfully to the ongoing advancement of biomechatronics and supports the development of the next generation of economical, smart, and user-friendly assistive devices (Brooker 2012).

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Ethical standard

The authors have no relevant financial or non-financial interests to disclose.

Availability of data and material

The data that support the findings of this study are available from the corresponding author upon reasonable request.

Conflicts of interest

The authors declare that there is no conflict of interest regarding the publication of this paper.

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