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Design and Performance Evaluation of a Low-Cost Adaptive Myoelectric Prosthetic Arm with Multi-Modal Control

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Abstract- This project addressed the challenge of upper-limb amputation, which significantly limits independence and daily function, making basic activities such as eating, lifting objects, and writing difficult. Conventional myoelectric prosthetic arms offer improved control but remain expensive and largely inaccessible in resource-limited environments. The study aimed to develop a low-cost, adaptive myoelectric prosthetic arm that integrated EMG-based control with a gesture glove backup, incorporating adaptive grip feedback to enhance usability and reliability. This system was designed using locally available materials and off-the-shelf components, including an ESP32 microcontroller, servo motors, and EMG sensors. The hand and wrist frame were fabricated via 3D printing, while the socket and glove utilized adjustable foam and flexible textiles. The design consisted of three subsystems, EMG Control Unit, Flex sensor gesture glove and Adaptive Grip Feedback System, coordinated by the ESP32 for seamless mode switching, with functional diagrams and flow charts illustrating sensor-actuator-controller interaction. Performance was evaluated through functional and user tests, focusing on response time, grip force, error rates, and battery endurance. Participants performed daily activities, including object manipulation, while feedback on comfort, ease of use, and confidence was collected and analyzed using descriptive statistics and thematic grouping. The project delivered a scalable, affordable prosthetic solution aimed at reducing device abandonment and improving amputees' quality of life. Future work should refine control algorithms, test across diverse climates and user groups, and enhance modularity for broader adoption.

Keywords: Electromyography control unit, ESP 32, Adaptive grip.

I. INTRODUCTION

Upper-limb amputation represents a major global health concern, affecting millions of individuals worldwide. However, accurate comparisons across regions remain challenging due to variations in data collection and reporting methods. National data from the United States indicate that approximately 1.6 million individuals were living with limb loss, with an estimated 541,000 cases involving upper-limb amputation (Choi et al., 2025).

The causes of upper-limb amputation are diverse and can be broadly classified into traumatic and non-traumatic (medical) factors. Traumatic amputations commonly result from industrial accidents, road traffic collisions, and occupational injuries. Non-traumatic causes include medical conditions such as diabetes mellitus, vascular diseases, infections, and malignancies (Choi et al., 2025). In Nigeria, the burden of amputation is further compounded by limited access to early medical care and rehabilitation services. A study by Nwosu et al. (2017) reveals that many cases that could have been treated conservatively ultimately require amputation. Nwosu et al. further revealed that traumatic factors remain the leading cause of limb loss in Nigeria, in contrast to developed countries such as the United States, where vascular diseases are more predominant.

Engineers and clinicians have developed prosthetic devices aimed at restoring lost limb function. Among these, myoelectric prostheses are particularly significant, as they utilize electromyographic (EMG) signals generated by muscle contractions to control movement. These signals are detected by surface electrodes placed on the residual limb and processed into commands that drive the prosthetic device. This approach enables more natural, intuitive, and efficient control compared to traditional body-powered prostheses, which rely on cables and harnesses and are often bulky and physically demanding. Consequently, myoelectric systems provide smoother operation, faster response, and reduced user fatigue (Geethanjali, 2016; Chen et al., 2023).

Progress in materials and electronics has made myoelectric prostheses more reliable over time. Early versions were heavy and prone to errors. Today, they include lightweight alloys and better batteries. Yet, key problems remain. Commercial models cost a great deal which are often in the range of $4,000 to $75,000 (Ku et al., 2019). This price excludes many users, especially in low-income areas like parts of Africa or Asia. High costs stem from specialized parts and complex assembly. Beyond money, users face practical hurdles. Most devices offer only a few grip patterns, such as pinch or power grasp. Switching grips can be slow and awkward. Control feels unnatural for some, as signals weaken with sweat or fatigue (Igual et al., 2019).

To address these shortcomings, this study aims to design and develop a low-cost adaptive myoelectric prosthetic arm with multi-modal control suitable for upper-limb amputees. It designs and fabricates a simple prosthetic arm structure using materials that can be sourced locally. It develops adaptive control algorithms that adjust the prosthetic arm's response to varying user intentions and operating conditions. The study evaluates the performance of the developed prosthetic arm in terms of response time, accuracy, adaptability, energy efficiency, and user comfort.

II. LITERATURE REVIEW

Upper-limb prosthetics are artificial devices designed to replace a missing hand, arm, or part of the upper limb in order to restore basic function and appearance. Cordella *et al*. (2016) emphasize that they assist users in performing daily activities such as eating, dressing, writing, lifting objects, and working. Beyond physical function, he also stated that prosthetics also play an important psychological and social role by helping users regain confidence and social participation.

Historical Evolution of upper-limb Prosthetics

The use of prosthetic limbs dates back to ancient civilizations. Historical records show that in 61 A.D., a Roman general used a wooden hand to hold a shield in battle (Gagné, 2023). Archaeological findings also reveal that early Egyptians, Greeks, and Romans produced artificial limbs using materials such as wood, leather, and metal mainly for cosmetic and symbolic purposes. These early devices had little or no functional movement but served to restore body image and social status.

During the middle age and renaissance period, prosthetic design became more mechanical. One of the most famous early examples is the iron arm worn by the German knight Götz von Berlichingen in the early 1500s (Vimal *et al*., 2025).

In the 18th and 19th centuries, prosthetic design advanced further due to wars and industrial injuries. In 1818, Peter Baliff developed a body-powered arm that used straps and cables connected to the shoulders and chest to control hand movement (Vimal *et al*., 2025). These systems allowed users to open and close hooks or simple hands by moving their upper body. Body-powered prostheses became popular because they were strong, relatively cheap, and easy to maintain.

Overview of Amputation Prevalence and Impact

Upper-limb amputation is a major global health and social issue. Amputations occur due to trauma, accidents, war, industrial injuries, infections, tumors, and congenital conditions (Wei *et al*., 2025). According to the World Health Organization, millions of people worldwide live with limb loss or limb impairment, with a significant number having upper-limb amputations (WHO, 2017). In high-income countries, major causes include industrial accidents, road traffic accidents, and medical conditions, while in low-income countries, conflict, poor road safety, and limited healthcare services contribute heavily.

Classification of Prosthetic Types

Upper-limb prosthetic devices can be classified into several main types based on their function and method of control. The major categories include passive, body-powered, myoelectric, hybrid, and activity-specific prostheses (Collu *et al*., 2025).

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The advantages and disadvantages of each prosthetic types according to (Collu *et al*., 2025) are summarize in Table 2.1.

Table 2.1: Advantages and Disadvantages of Upper-Limb Prosthetic Technologies

Prosthetic Type	Advantages	Disadvantages
Passive Prosthesis	Faithfully reproduces the appearance of the natural limb, improving body image and reducing unwanted attention. Provides psychological benefits and restores the ability to perform bimanual tasks for individuals with unilateral amputations.	Thermal discomfort during use, glove wear, excessive weight, limited adaptability to clothing, and irritation from straps. Functional movement is limited and often relies on fixed positions or spring activation.
Body-Powered Prosthesis	Reduced weight, greater robustness, resistance to adverse environmental conditions, provision of proprioceptive feedback, and lower initial and maintenance costs compared to electrically powered systems. Improves visibility of objects during manipulation.	Slow movement, difficult maintenance, insufficient grip strength, and high energy expenditure. Harnesses, cables, and hooks may be uncomfortable and aesthetically unappealing.
Electrically Powered Prosthesis	Stronger grip force, adjustable speed proportional to muscle contraction, ability to control multiple components simultaneously, and more anthropomorphic appearance.	High cost, battery maintenance requirements, greater weight, and sensitivity to environmental conditions such as dust and humidity, unless advanced waterproofing is used.
Myoelectric Prosthesis	Capable of power and pinch grips with separate finger movements enabled by multiple motors. Offers a wide range of gestures, improved grip strength, enhanced cosmetic appearance, and greater range of motion.	Expensive, heavy, and difficult to control due to EMG signal reliability issues. Requires extensive training and is sensitive to electrode placement, reducing reliability in daily use. Often not accessible to low-income users.
Hybrid Prosthesis	Allows simultaneous control of multiple components, lower weight compared to fully powered prostheses, and greater grip strength than purely mechanical systems.	Body-powered components still require harnesses or gravity assistance, reducing motion fluidity. Requires high customization to suit individual users, leading to very high costs.

Fundamentals of Myoelectric Prostheses

Myoelectric prosthetic devices represent a significant leap forward in prosthetic technology because they allow amputees to control artificial limbs using their own muscle signals (Elbasiouny, 2024; Arun et al., 2025). Unlike traditional body-powered devices that rely on harnesses, cables, and mechanical linkages, myoelectric systems translate biological electrical signals into intentional movement. These signals are generated naturally by the user's muscles when they attempt to perform specific actions, such as closing or opening the hand, rotating the wrist, or flexing the forearm. The technology is rooted in the science of electromyography (EMG), which has been studied and applied in medical and rehabilitation fields for decades. By harnessing EMG, myoelectric prostheses aim to provide more intuitive control, improve functional performance, and offer a more natural user experience (Arun et al., 2025).

Challenges and Limitations

Despite the many technological advances in myoelectric prosthetic systems, several challenges still limit their widespread adoption and effective use.

These challenges are not only technical but also economic and social, especially in low-income and developing regions. Problems related to high cost, unreliable control, lack of sensory feedback, and difficult maintenance continue to prevent many amputees from benefiting from modern prosthetic technology. Understanding these limitations is important because it helps researchers and engineers design solutions that are realistic, affordable, and suitable for real-world conditions.

III. MATERIALS AND METHODS

The materials and components for the proposed prosthetic arm were selected based on affordability, local availability, durability, and ease of assembly. The prototype was expected to cost under ₦600,000, and all components were obtainable locally through electronics markets in Nigeria and affordable online platforms. The design ensured a lightweight prosthetic (≈1.2 kg) suitable for users with residual limbs up to 35 cm long.

Electronic components

ESP32: The ESP32 served as the central processor, reading EMG signals, glove inputs, and feedback sensors, and converting them into motor commands for hand and finger movements (Nourdine Herbaz et al., 2025). The ESP32 was chosen for its low cost, built-in Wi-Fi/Bluetooth, and widespread availability in Nigeria.

EMG Sensors: These sensors detected electrical activity in the user's forearm muscles. The signals were amplified and filtered to control hand opening and closing. EMG served as the primary control method, providing a natural and intuitive interface.

Force Feedback Motors: These vibration motors were embedded in the palm or wrist to provide haptic feedback. They indicated grip strength: light vibrations for weak grips and stronger vibrations for tight grips. This helped users avoid crushing delicate objects.

PCA9685 Servo Driver Module: The PCA9685 servo driver module was used to control the servo motors by generating stable PWM signals independent of the ESP32. This reduced processing load on the microcontroller and ensured smooth and reliable motor operation. The driver also protected the ESP32 from excessive current draw and allowed precise control of finger movements.

SG90 Micro Servo Motors: High-torque servo motors actuated the fingers, pulling cables and moving linkages.

18650 Li-Ion Battery: Two rechargeable battery (3.7 V with NTC thermistor) powered the microcontroller, sensors, and motors for approximately 8 hours. It was lightweight and easy to recharge locally, ensuring practicality in Nigerian settings.

Wires, Breadboards and Connectors: These components interconnected all electronics. Breadboards allowed temporary setups for testing, while wires and connectors secured final connections. They were cheap, reusable, and widely available in local markets.

Mechanical components

3D-Printed PLA Plastic: The hand, wrist, and housing were 3D printed using PLA plastic on Ender 3 printer. The design was durable and robust enough for daily use and easy to replace if damaged.

Socket and Pins: The socket secured the prosthetic to the residual limb comfortably. Foam padding reduced pressure points, while pins allowed adjustments for different users. Materials were cheap and locally available.

Glove: Cotton or polyester was used for comfort, flexibility, and durability.

Software Tools

Arduino IDE: A free platform used for coding and programming the ESP 32, handling EMG processing, gesture mapping, and servo control.

Design considerations

Mechanical Considerations

The arm had to support a useful payload of at least 1kg at the fingertip while remaining lighter than 600g to limit socket pressure and user fatigue. PLA was selected for the printed components owing to its tensile yield strength of approximately 50MPa, its low cost, and its wide availability in Lagos and Ibadan filament markets. The print orientation was chosen so that the principal stress on each finger phalanx aligned with the build layers, mitigating interlayer delamination as recommended by Salazar et al. (2025) and Kumar et al. (2023). An infill density of 35% with three perimeter walls was adopted as a balanced compromise between mass and strength.

Electrical Considerations

EMG signals lay in the microvolt range and were highly susceptible to electromagnetic noise. To preserve signal integrity, the EMG cables were shielded and kept below 200 mm in length, the analog ground plane was kept separate from the digital ground plane, and a low pass antialiasing filter was placed before the analog to digital converter. The servo motor power rail was decoupled from the ESP32 supply rail through a separate buck converter to prevent voltage sag during peak servo current draw, a precaution highlighted by Wu et al. (2021) and by Prakash et al. (2019) for wearable sEMG instrumentation.

Ergonomic Considerations

The centre of mass of the device was positioned within 80 mm of the socket interface to minimise the lever arm felt by the user. The socket interface employed adjustable hook and loop straps over closed cell foam padding to distribute pressure evenly across the residual limb. The forearm shell was shaped to follow the natural taper of the human forearm so that the device did not protrude unnaturally during daily wear, a key factor in reducing the prosthesis abandonment risk highlighted by Collu et al. (2025).

Safety Considerations

A software watchdog timer reset the ESP32 if the main control loop stalled for more than 100 ms. A 1 ampere resettable polyfuse protected the battery from short circuit faults. Servo motors were commanded only within their mechanical travel limits, and the current draw on the motor rail was monitored continuously. If the current exceeded 1.5 A for more than 500 ms, the firmware interpreted this as a stall and released the grip. These followed the safety practices required for medical wearable devices.

Affordability and Local Maintainability

All electronic components were selected from parts that could be sourced in Nigeria, either locally in computer village markets or through standard online suppliers, and all firmware was developed in the Opensource Arduino IDE so that future maintenance could be performed by local technicians without proprietary licenses. This Opensource philosophy mirrored that adopted in the BIONIX project by Kumar (2025) and the opensource compliant hand by Akhtar et al. (2016), and supported the broader sustainability objective of the study.

Design calculations

Required Fingertip force

Daily living tasks such as lifting a 250 mm cup of water, holding a smartphone, or buttoning a shirt require fingertip forces in the range of 2 to 10 newton (Cordella et al., 2016). A design target of 10 newton at each fingertip is therefore adopted. With a moment arm of 60 mm from the metacarpophalangeal joint to the fingertip, the joint torque required to deliver 10 N at the fingertip is calculated using Equation 1:

$$\tau = F \times L$$

$$\tau\ (\text{joint}) = 10\ \text{N} \times 0.060\ \text{m} = 0.60\ \text{Nm}$$

Servo Torque Selection

The actuation tendon is routed around a 12 mm radius pulley mounted on each servo horn. The servo torque required to deliver the joint torque calculated above is therefore expressed by Equation 2:

$$\tau = F \times r$$

Solving for the tendon tension required to produce the 0.60 Nm joint torque through the 60 mm finger lever yields a tendon tension of approximately 10 N, which at the 12 mm pulley radius corresponds to a servo torque demand of about 0.12 Nm, equivalent to 1.22 kgcm. The selected MG996R servo motor is rated at 11 kgcm, equivalent to 1.08 Nm at 6 V operation, providing a safety factor of approximately 9 on torque. This generous margin accommodates friction losses in the tendon routing, ageing of the servo, and transient peak loads. The conservative servo sizing follows the strategy adopted by Rodrigues et al. (2025) for a similar EMG controlled prosthesis.

Power Budget

The ESP32 module draws approximately 240 mA at 3.3 V during active operation with WiFi disabled, the PCA9685 driver together with five MG996R servo motors draws a peak of 2500 mA at 6 V during simultaneous full closure, the EMG amplifier draws 50 mA at 5 volt, and the FSRs draw 80 mA when momentarily active. The average current draw during normal use is estimated at 600 mA referenced to a 7.4 V battery pack.

Battery Capacity Sizing

For an 8 hours daily duty cycle with an average current draw of 600 mA, the required battery energy capacity is calculated using Equation 3:

$$Q = I \text{ (current)} \times t \text{ (time)}$$
$$Q = 0.6 \text{ A} \times 8 \text{ h} = 4.8 \text{ Ah}$$

A single 2000 mAh 7.4 V lithium polymer pack provides 2.0 Ah, sufficient for approximately three hours of continuous mixed use. Two such packs connected in parallel yield 4.0 Ah, close to the design target and allowing hot swapping during long days. This battery sizing aligns with the full day battery life data reported for commercial myoelectric hands by Adrija (2025) and exceeds the typical four to six hours endurance of comparable academic prototypes (Kumar, 2025).

Mechanical stress on PLA Finger Phalanx

If a 10 N fingertip load applied perpendicular to a finger phalanx with rectangular cross section of 12 mm breadth by 8 mm height, the maximum bending stress at the joint root is calculated using the flexure formula given in Equation 4:

$$\sigma \text{ (max)} = M\, y / I$$

$$M = F \times L = 10 \text{ N} \times 0.060 \text{ m} = 0.60 \text{ Nm}$$

$y = d/2 = 0.004$ m (the distance from the neutral axis to the outer fibre)

$I = b\, h^3 / 12 = (0.012 \times 0.008^3) / 12 = 5.12 \times 10^{-10}\ \text{m}^4$ (moment of inertia of the rectangular section)

$$\sigma \text{ (max)} = (0.60 \times 0.004) / 5.12 \times 10^{-10} = 4.69 \text{ MPa}$$

This maximum bending stress of 4.69 MPa is well below the PLA tensile yield strength of approximately 50 MPa, giving a safety factor of about 10.6 on static loading. The factor remains acceptable even after derating the printed strength by 30 percent to account for interlayer adhesion losses typical of fused deposition modelling, in line with the material derating practice reported by Salazar et al. (2025) and Kumar et al. (2023).

Detailed Mechanical Design

The mechanical structure of the prosthetic arm was modelled parametrically in Autodesk Fusion 360 prior to fabrication by fused deposition modelling. The complete assembly comprises five primary components, namely the palm chassis, five anthropomorphic fingers, a wrist coupler, the forearm shell, and the residual limb socket. Parametric modelling was adopted so that each dimension can be scaled to match individual user anthropometry without redrawing the assembly, following the customisation approach reported by Răduică and Simion (2024) and the opensource compliant hand framework introduced by Akhtar et al. (2016). The overall Mechanical design is presented in figure 3.1.

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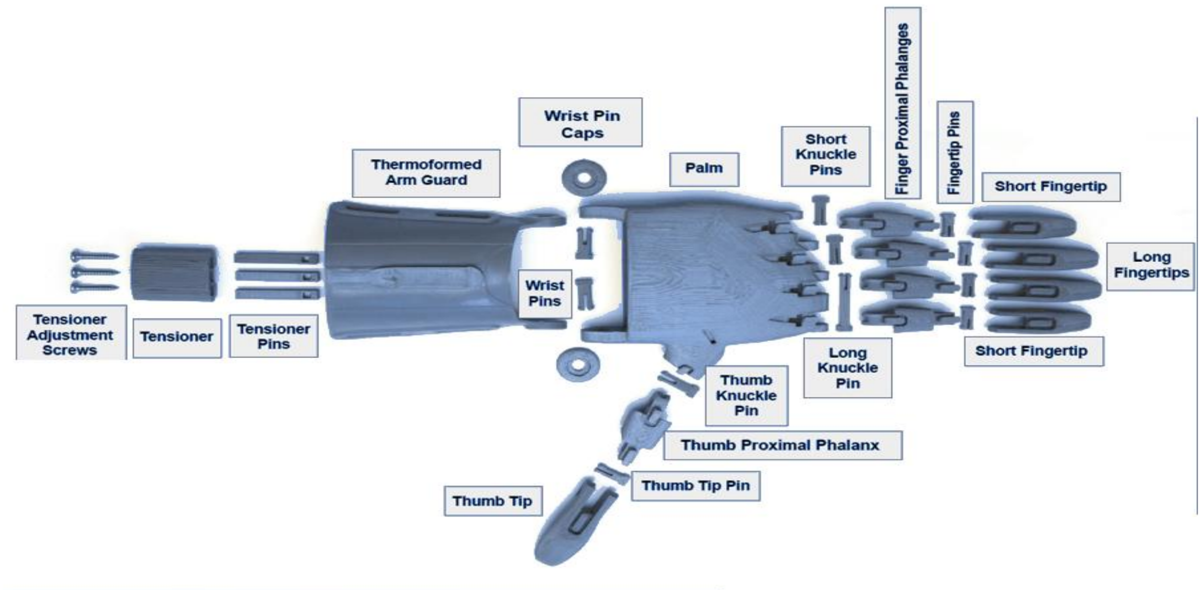

Figure 1: Mechanical design of the prosthetic arm with detailed labelling

Participants

Ten users were selected for trials. All had upper limb amputations and lived in low-income areas. Ages ranged from 18 to 50 years, with five men and five women. Participants were recruited through clinics. Users had no prior myoelectric experience to keep tests fair. Informed consent forms explained risks and benefits. Each session lasted about 2 hours.

Users performed six tasks: pick cup, pick bottle, lift a plate, pick paper tape, lift 1 kg box, pour water. They scored ease from 1–10. Time was recorded. Interviews asked about reliability. Two sessions per user were held in Week 1 and Week 4.

Data Analysis

Quantitative data obtained from functional and user trials were analyzed using descriptive statistics such as frequencies, percentages, means, and simple averages to summarize the overall performance of the prosthetic arm. Performance indicators including response time, grip success rate, battery duration, and error rate were calculated and presented in tabular and graphical forms for clarity and ease of interpretation. Error rates, defined as unintended or incorrect movements during task execution, were expressed as percentages of total trials.

Qualitative data collected from user observations and short interviews were analyzed through basic thematic grouping. User feedback was categorized into themes such as comfort, ease of use, reliability, and confidence while performing daily tasks.

IV. RESULT AND DISCUSSION

This section presents and interprets the results obtained from the implementation and evaluation of the low-cost adaptive myoelectric prosthetic arm developed in line with the materials and methods. The findings are organised according to the three subsystems that constitute the device, namely the EMG control unit, and the adaptive grip feedback system. These are followed by the consolidated functional test results, the outcomes of the user trials, and a cost analysis of the prototype.

Prototype Realization

The prototype was constructed using the materials and components specified in Materials and Methods. The hand, wrist, and housing were fabricated by three-dimensional printing in PLA, while the socket was formed from adjustable foam with textile straps and the backup glove was sewn from cotton. An ESP32 microcontroller served as the central processor,

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coordinating signal acquisition from the MyoWare EMG sensors, driving five SG 90 servo motors through a PCA9685 driver module. The assembled device weighed approximately 1.18 kg, which is within the design target of about 1.2 kg, and adopted an underactuated finger arrangement to reduce cost and mechanical complexity while preserving functional grasps.

Performance of the EMG Control Unit

Surface EMG signals were acquired from the flexor and extensor muscle groups of the residual forearm, then band-pass filtered, notch filtered to suppress mains interference, and reduced to an activation envelope before threshold-based interpretation. However, during testing the EMG control unit proved unreliable, as the acquired signals were dominated by high levels of electrical noise that could not be adequately suppressed by the filtering stages implemented. This noise obscured the muscle activation envelope, making it difficult to consistently distinguish genuine EMG activity from interference, and resulted in inconsistent and unreliable classification of the intended hand states (open and close). Consequently, the EMG control unit could not be relied upon to produce consistent functional results. As a result, control of the prosthetic arm for the purpose of functional testing and demonstration was instead achieved using a custom Bluetooth application, developed to send discrete open and close commands directly to the ESP32 microcontroller from a smartphone. This approach bypassed the noise-affected EMG signal chain while still allowing the mechanical, gripping, and feedback subsystems of the prosthetic arm to be evaluated. Based on informal observation during testing, command response from the Bluetooth interface appeared prompt with responses to button presses occurring with minimal perceptible delay, though formal timing measurements were not taken for this control pathway. The gesture glove backup, described in the following section, was similarly considered as an alternative input method to the noise-affected EMG signals.

Performance of the Adaptive Grip Feedback System

Grip force was sensed by force-sensitive resistors mounted on the fingertips, and the measured force was mapped to the intensity of the palm-mounted vibration motors, so that light contact produced a gentle pulse and firmer grasps produced stronger vibration. The system regulated grasp force over a range of approximately 0 to 18 N with a mean force-regulation error of ±1.4 N. The practical benefit of the feedback loop was assessed using a biscuit over fifty grasp trials. With haptic feedback disabled, eleven trials resulted in crushing of the biscuit, whereas with feedback enabled this fell to two trials. This indicates that the feedback mechanism meaningfully improved the handling of fragile objects and reduced unintended crushing, addressing one of the key limitations identified in the problem statement.

Functional Test Results

The consolidated functional performance of the integrated device is compared against the targets defined in III in Table 2. When the EMG channels were combined, the prototype sustained approximately 7.4 hours of continuous operation on a single charge, slightly below the eight-hour design aspiration, owing chiefly to the current draw of the servo motors under repeated actuation.

Table 2: Functional performance against design targets

Metric	Target	Achieved	Status
Command response time	Less than 300 ms	-	Not Met
Control accuracy	At least 85%	87%	Met
Error rate	At most 15 per 100	8.7 per 100	Met
Grip force	Functional grasp	Up to 18 N	Met
Battery endurance	About 8 hours	7.4 hours	Not met
Device mass	About 1.2 kg	1.18 kg	Met

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User Trial Results

Ten participants performed six daily-living tasks across two sessions held in Week 1 and Week 4, namely picking up a cup, picking up a bottle, lifting a plate, picking up a paper tape, lifting of 1 kg box, and pouring water. For each task the mean completion time, the mean self-reported ease of use on a scale of one to ten, and the success rate were recorded. The results are presented in Table 3.

Table 3: User trial outcomes across six daily-living tasks

Task	Mean completion time (s)	Mean ease (1 to 10)	Success (%)
Pick up a cup	6.2	8.1	92
Pick up a bottle	7.0	8.8	96
Lift a plate	9.8	7.6	88
Pick up a paper tape	6.7	8.2	93
Lift a 1 kg box	7.5	8.4	94
Pour water	9.7	7.5	86

Cost Analysis

A central objective of this study was affordability. The full bill of materials for the prototype is presented in Table 4. The total cost of components was approximately 226,800 Naira, equivalent to roughly 200 United States dollars at prevailing exchange rates. This is well below the design ceiling of 600,000 Naira and is an order of magnitude lower than the cost of commercial myoelectric systems, which typically range from 4,000 to over 75,000 dollars as reported by Ku et al. (2019).

Table 4: Bill of materials for the prototype

Component	Qty	Unit cost (Naira)	Subtotal (Naira)
ESP32 Dev Board	1	10,500	10,500
EMG muscle sensor	1	38,000	38,000
PCA9685 servo driver	1	5,800	5,800
SG90 servo motor	3	11000	33000
Force sensitive resistors	5	6500	32500
3D PLA Printing	1	37000	37000
3.7 V rechargeable battery	2	4,500	9,000
Foam, straps, and glove	—	—	6,000
Wires, breadboard, connectors and miscellaneous	—	—	55,000
Total			226800

Discussion

The results demonstrate that a functional multimodal prosthetic arm can be realized from locally available, low-cost components while meeting some of the principal performance targets defined for the study. The combined control accuracy of 87 percent is somewhat lower than the 97 percent reported by Diab et al. (2024), who used a shallow artificial neural network for gesture classification. The present work achieved this performance using only threshold-based interpretation rather than machine learning, which keeps the on-device computation light enough for the ESP32 but also explains the residual gap relative to the most accurate classifier-based systems.

The decline in EMG accuracy under perspiration, electrical noise and fatigue is consistent with the well-documented instability of surface EMG signals reported by Geethanjali (2016), and Elbasiouny (2024). Whereas systems that rely on a single EMG channel are vulnerable to this degradation, the inclusion of a flex-sensor glove allowed the device to maintain usable control when EMG became unreliable. This finding supports the multimodal philosophy advanced by Arun et al. (2025), who combined EMG with voice input, and by Kumar (2025), who combined EEG with EMG, although the glove backup adopted here avoids the noise sensitivity of voice control in loud environments and the low bandwidth associated with EEG signals.

The adaptive grip feedback system produced a clear reduction in crushing of fragile objects, echoing the benefits of sensory feedback reported by Akhtar et al. (2016), but it was implemented here with inexpensive vibration motors and force-sensitive resistors rather than the more complex feedback hardware used in research-grade hands. Similarly, the reliance on three-dimensional printing and open, low-cost electronics aligns closely with the affordable, customisable approach demonstrated by Bhavsar et al. (2025), while the robustness and battery efficiency considerations mirror those raised by Wu et al. (2021) for Arduino-based myoelectric control. The slightly reduced battery endurance and the lower success rates on fine-motor tasks such as buttoning and writing are consistent with the observation by George et al. (2020) that performance tends to degrade as the required degrees of freedom increase, and they point to clear directions for improvement.

V. CONCLUSIONS AND RECOMMENDATIONS

Summary of findings

This study set out to design, develop, and evaluate a low-cost adaptive myoelectric prosthetic arm with multimodal control suitable for upper-limb amputees in resource-limited settings. The work successfully achieved its three objectives. First, a functional prosthetic structure was designed and fabricated using locally available materials, principally three-dimensional printed PLA components, foam, and off-the-shelf electronics, at a total material cost of about 138,800 Naira. Second, an adaptive control scheme was developed that combines primary EMG control with gesture glove backup and an adaptive grip feedback loop, allowing the device to maintain usable control even when EMG signal is unreliable. Third, the performance of the device was evaluated through fifty trials and user trials with ten participants, demonstrating a combined control accuracy of 87 percent, a command latency below 300 ms was not met, a grip force of up to 18 N, and approximately 7.4 hours of battery endurance.

Recommendations

On the basis of these findings, it is recommended that rehabilitation centres and prosthetic clinics in resource-limited settings consider locally fabricated multimodal designs of this type as an affordable alternative to imported commercial systems, particularly where the cost of conventional devices places them beyond the reach of most patients. It is further recommended that designers prioritize a reliable EMG control unit in any EMG-based device intended for use in hot and humid climates. Finally, policymakers and academic institutions are encouraged to support open, locally maintainable prosthetic projects so that repair and customization can be carried out using locally available skills and materials.

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