Physical and Engineering Sciences in Medicine (2021) 44:229–241
https://doi.org/10.1007/s13246-021-00972-w

SCIENTIFIC PAPER

# A low-cost transradial prosthesis controlled by the intention of muscular contraction

**Alok Prakash**[1] · **Shiru Sharma**[1]

Received: 30 July 2020 / Accepted: 7 January 2021 / Published online: 19 January 2021


## Abstract
Persons with upper-limb amputations face severe problems due to a reduction in their ability to perform the activities of daily living. The prosthesis controlled by electromyography (EMG) or other signals from sensors, switches, accelerometers, etc., can somewhat regain the lost capability of such individuals. However, there are several issues with these prostheses, such as expensive cost, limited functionality, unnatural control, slow operating speed, complexity, heavyweight, large size, etc. This paper proposes an affordable transradial prosthesis, controlled by the muscular contractions from user intention. A surface EMG sensor was explicitly fabricated for capturing the muscle contraction information from the residual forearm of subjects with amputation. An under actuated 3D printed hand was developed with a prosthetic socket assembly to attach the remaining upper-limb of such subjects. The hand integrates an intuitive closed-loop control system that receives reference input from the designed sensor and feedback input from a force sensor installed at the thumb tip. The performance of the EMG sensor was compared with that of a traditional sensor in detecting muscle contractions from the subjects. The designed sensor showed a good correlation ($r > 0.93$) and a better signal-to-noise ratio (SNR) feature to the conventional sensor. Further, a successful trial of the developed hand prosthesis was made on five different subjects with transradial amputation. The users wearing the hand prototype were able to perform faster and delicate grasping of various objects. The implemented control system allowed the prosthesis users to control the grasp force of hand fingers with their intention of muscular contractions.

**Keywords** Electromyography · Transradial prosthesis · Control system · 3D printing · Signal-to-noise ratio

## Introduction

Upper-limb amputation has an immense effect on the patient's body image and working capability. The majority of upper-limb losses are transradial, i.e., below-elbow, which are mostly reported from developing countries [1, 2]. Accidents, malignant tumours in bone or muscles, infections, etc., are the chief reasons responsible for such amputations [3]. Prosthetic arms are considered the primary solution to below-elbow amputations, as these are capable of restoring some functions of the missing limb. Based on functional capability, these are classified as body-powered and externally-powered prostheses [4]. The body-powered prosthesis consists of hooks, which are controlled by cable. Initially, the hook remains closed; when body power pulls the cable through the harness, the hook opens to grasp the object. The main disadvantage of the body-powered prosthesis is it requires considerable body power to actuate the prosthesis, produces unnatural grasping for objects, and creates discomfort while wearing [5].

Externally powered prostheses, on the other hand, are battery operated and are controlled by bioelectric signals like electromyography (EMG), electroencephalography (EEG), electrocorticography (ECOG), and from other electrical signals generated by sensors and switches [6, 7]. EMG and EEG techniques are mostly preferred for controlling prostheses because of their non-invasiveness, whereas ECOG and TMR are invasive techniques [8, 9]. EEG can be a decent option for directly controlling the prosthetics with brain signals. However, there are several issues with these signals such as low reliability, low accuracy, low user adaptability, low data transfer rate, and complex acquisition setup [10].

A myoelectric prosthesis is a type of externally powered prosthesis, which utilizes EMG signals from the residual upper-limb of patients to operate the terminal device, using

✉ Alok Prakash
alokp.rs.bme15@itbhu.ac.in

1 School of Biomedical Engineering, Indian Institute of Technology (BHU), Varanasi 221005, India

a suitable control system [11, 12]. EMG can directly reflect the motion intention of the user through muscular contractions, which can be utilized to control prosthetics intuitively. Moreover, EMG requires easy acquisition setup and are reliable as compared to other means [13–15]. Nowadays, EMG has become a prominent source of control for the upper-limb prosthesis [16–18].

The control system used for hand prosthesis mostly consists of an upper-level control or control strategy for translating the input EMG signal to control instruction and low-level control for generating final output based on the force, velocity, or position feedback [19, 20]. The feedback can be supplied using different tactile sensors that are present within the prosthetic devices. To achieve precise and delicate grasping of objects with the hand, low-level control is used along with upper-level control [21, 22].

In the last few years, there is tremendous progress regarding the development of biomimetic hand prosthesis. Substantial researches have been performed, resulting in the evolution of anthropomorphic myoelectric hands, which can offer features such as multiple grip patterns, precise grasping, and individual finger movement [23–27]. The operation of these prosthetic devices is based on a pattern recognition system receiving inputs from single or multi-channel EMG devices or other devices like pressure sensors, inertial measurement units (IMU), Hall effect sensors, etc. However, most of these products are still limited to research works only, i.e., their clinical applicability is yet to be realized. The reason behind the unacceptability of these prosthetic devices among users is expensive cost, large size as well as weight, complexity, unnatural control, speed of operation, etc.

Bebionic v3, Michelangelo, Vincent evolution 3, and I-limb quantum are some commercialized prosthetic hands having great features such as multiple grip patterns, precise grasping, lightweight, individual finger operation, etc., which can restore the lost functionality of upper-limb amputees [28–31]. However, these devices are too much expensive for the amputees, which belong to low-income countries. Moreover, these hands are based on sophisticated control systems, which require enormous pieces of training for their reliable operation [32–34]. The survey report reveals that more than 85% of persons with transradial amputations belonging to developing countries cannot afford functional prosthetics [2, 35, 36]. The majority of these patients are still using either cosmetic or body-powered prostheses, which cannot fulfill their basic life needs.

Patients with amputation require a simple, affordable, fast, lightweight, robust, and dexterous hand that can perform activities of daily livings (ADLs) with minimum training efforts. Increasing the number of features may enhance hand prosthesis functionality, but this can also increase the cost, complexity, and weight of the prosthesis [24, 37, 38]. Therefore, while designing a prosthetic hand, all the parameters should be considered carefully such that it becomes affordable and functional enough to accomplish the prosthesis user's basic requirements.

This work presents a low-cost transradial prosthesis controlled by muscular contractions from the user's intention. A specific wearable EMG sensor was designed for the reliable detection of muscle activity from the residual upper-limb of amputees. The sensor's ability to capture the EMG signal was compared with that of a commercial sensor. A closed-loop based position control system was formulated and implemented in real-time for an indigenously prepared 3D printed hand prosthesis. In the control system, the designed sensor produced the reference input, whereas a force sensor installed at the thumb tip of the developed hand provided feedback. Further, the developed hand prosthesis was tested on below-elbow amputees for executing various grasping tasks.

## Materials and methods

### EMG sensor

Surface electromyography (sEMG) is the non-invasive approach for the measurement of the electrical activity of muscle. The bioelectric signal measured by this technique has a magnitude of $\leq 10$ mVp-p, which consists of significant information in the frequency range of 0–500 Hz. Surface EMG can be applied to estimate the intention, force, limb angle, and level of muscle contraction of the subject [39–42]. These features are utilized in several applications, including the study of neuromuscular disorders, muscle fatigue analysis, and control of assistive devices [16–18].

This research mainly emphasizes the development of myoelectric hand prosthesis, which requires a low-cost, wearable sEMG device capable of providing a good-quality signal with an open-source feature. As the commercial research-grade, devices can be expensive and are unable to provide an open-source platform [43, 44]. Therefore, in this work, an explicit EMG sensor was designed for the application of prosthesis control.

#### Sensor design

The sEMG sensor was designed using three-electrode connectors, a pre-processing hardware section, and a power supply unit, all integrated into a single structure. The block diagram of the sensor describing its various parts is shown in Fig. 1.

Surface electrodes provide the crude assessment of bioelectric signals under the skin. Stainless steel buttons were incorporated as the electrode connectors, which attach to the skin through Ag–AgCl electrodes for capturing raw EMG

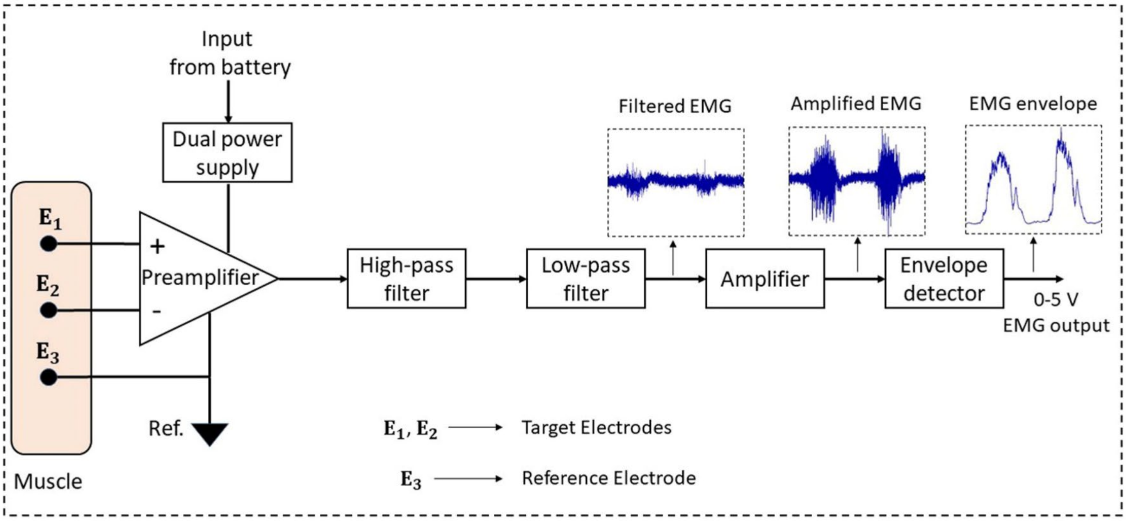


**Fig. 1** Block diagram of the proposed sensor

signals. The two target electrode connectors were mounted on the 3D printed sensor chassis at an inter-electrode distance of 3 cm, while the third connector was freely held for connecting the reference electrode [45]. Such a differential arrangement of electrodes diminishes the possibility of movement artifacts, which typically arise due to electrode cables' motion [46, 47]. Figure 2b depicts the rear view of the sensor showing the electrode connectors.

The pre-processing hardware section of the sensor was designed with a sequence of stages such as a preamplifier, a band pass filter, an amplifier, and an envelope detector.

An instrumentation amplifier (IC-128P) having features such as low offset voltage, high common-mode rejection ratio (CMRR), low power consumption, and adjustable gain was employed as a preamplifier stage. A band pass filter (i.e., a high pass filter followed by a low pass filter) of a bandwidth of 20–350 Hz was incorporated to filter out the low and high-frequency noise components from the raw EMG signal [48]. Sallen key type second order high and low pass filter circuits were employed for the band pass filter stage. The amplitude of the filtered EMG signal was further enhanced using a simple inverting amplifier. Until this stage, overall gain of the amplifiers was achieved around 3500. To generate the linear envelope of the filtered and amplified signal, an envelope detector stage consisting of a precision rectifier circuit and a low pass RC circuit was used [49]. Such a step makes the amplitude of the EMG signal unchanged by the other factors except for the contraction of muscles [50].

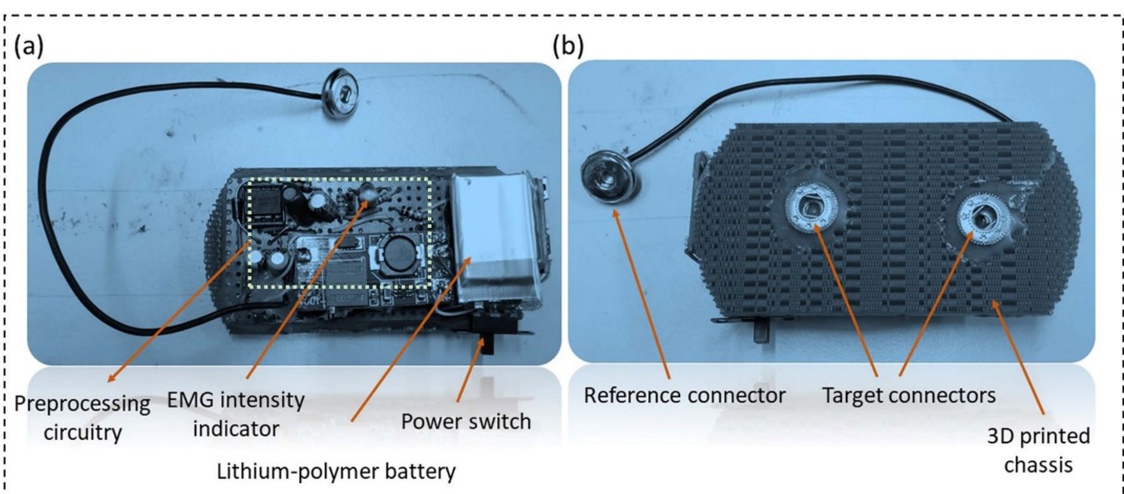


**Fig. 2** **a** Sensor's front view, **b** Sensor's rear view

The various components present within the preprocessing section was powered through a 3.7 V, 250 mAh rechargeable battery. A dual power supply circuit was used in series with a step-up booster circuit to produce a bipolar supply (from the battery) for powering active components [44].

### Sensor description

The sensor's front and rear views are described in Fig. 2a, b showing the various components of the sensor. The front side of the sensor mainly displays the preprocessing circuitry and the power supply. The final output produced by the sensor was a 0–5 V linear envelope proportional to the intensity of muscular contraction. A light-emitting diode (LED) on the front view indicates the strength of the captured EMG signal. To power on the device, a single pole double throw (SPDT) slider switch was employed. The power consumption of the sensor was estimated as 25 mA using the datasheets of the used components.

### Sensor validation

**Correlation with conventional sensor** The ability of the designed sensor in detecting muscular contractions from the forearm of subjects was compared with a traditional sensor (Myoware). Myoware muscle sensor from Sparkfun Electronics is a low-cost sensor, which has decent applications in prosthetics [17, 18, 51, 52]. Both the sensors were positioned at flexor carpi ulnaris muscles one after one, and acquisitions of EMG signals were performed for four different contractions: (1) maximum voluntary contraction (MVC), (b) 75% of MVC, (c) 50% of MVC and (d) 25% of MVC. The percentage of contraction was measured with the help of a handheld dynamometer (form AD instruments). The flexor muscles (i.e., carpi ulnaris and radialis) situated on the forearm are mainly responsible for flexion/extension of fingers and wrist [53]. Approval was taken from the Ethical Committee, Institute of Medical Sciences, BHU, Varanasi, before performing the experiment. Both the sensors were placed in contact with the skin via disposable Ag–AgCl electrodes. The EMG data were captured using a data acquisition (DAQ) device with a sampling rate of 1 kHz. A correlation analysis was performed between the recorded EMG signals from the two sensors.

**Signal-to-noise ratio** The noise performance of both the sensors was quantified in terms of signal-to-noise ratio (SNR). SNR was evaluated as the ratio of the root mean square (RMS) value of the signal for MVC to the RMS value of the signal for no contraction, i.e., baseline noise [54]. Figure 3 describes the waveform indicating the EMG signal and noise level for determining SNR. Utilizing each recorded data of 5 s duration, SNR values for both the sensors were computed for all the ten subjects (using Eq. 1).

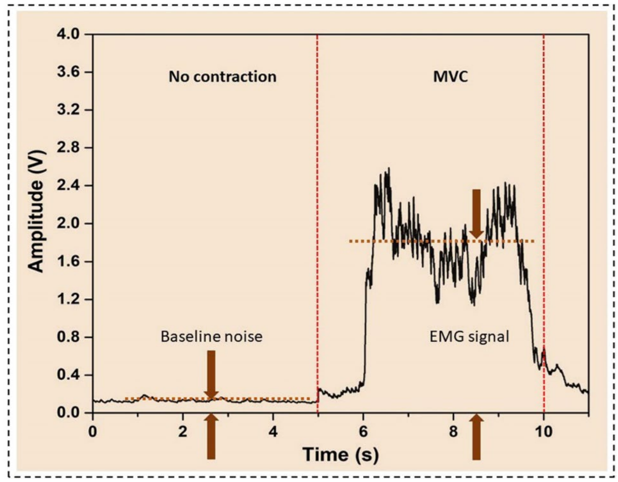


**Fig. 3** EMG waveform showing the signal and noise level

$$SNR = 20log_{10}\left(\frac{RMS_{signal}}{RMS_{noise}}\right) \quad (1)$$

## Development of hand prosthesis

### 3D printed hand parts

3D printing offers to manufacture low-cost, lightweight, decent strength, and customized products as per our design. A hand (along with socket) model was custom-made and was 3D printed using fused deposition modelling (FDM), employing polylactic acid (PLA) filament of 1.75 mm diameter[55]. The printing was performed with a filling density of 50% and the extruder temperature set at 220° C [56]. All the printed parts were assembled to form the hand prototype and its socket assembly. The socket consists of two pre-allocated space to fit servomotors.

### Actuation

The hand fingers were actuated using two digital servomotors (DS-3225) located on the prosthetic socket (i.e., away from the palm). Such a scheme is referred to as extrinsic actuation, which provides a more natural and biomimetic weight distribution to hand [57]. Flexion of hand fingers was produced by motor-tendon based actuation, whereas extension was provided by elastic elements attached to the phalangeal joints. Any angular displacement to the servomotor (through pulley) pulls the tendons to flex the fingers. Such a method translates the high torque of the servomotor to amplified linear flexion force without any loss of speed [27]. Figure 4a describes the primary actuation method for the hand fingers. Out of the two motors, one was used to actuate the thumb and the index finger while the other motor actuated the rest fingers. The hand was able to provide a total of two degrees-of-freedom (DOF) since each finger had two joints. Such a mechanism is underactuated, where the numbers of actuators are ≤ the number of DOF [27]. The servomotor accepts PWM input to produce angular displacement in the range of 0–180$^0$ and maximum torque up to 25 kgf cm. High tension fishing line of 0.6 mm was incorporated as a tendon.

### Force sensor

A force-sensitive resistor (FSR) was installed at the thumb tip to monitor the contact force while finger-object interaction (i.e., grasping). To promote even distribution of grasp force over the contact surface area, FSR was sandwiched between two thin layers of polydimethylsiloxane (PDMS) material [58]. Figure 4b shows the installation of a force sensor at the tip of the thumb. A simple voltage divider circuit was incorporated for translating the change in resistance to the output voltage. The output of the force sensor in 0–5 V serves as a feedback signal providing information about the prehension force during grasping. FSR is a low-cost and optimum sensor for measuring the force over the current sensors in prosthesis applications [59].

### Control system

The Control system in a myoelectric prosthesis mainly decides how the actuation of fingers will be done in accordance with the input EMG signal [33]. The proposed control system in Fig. 5 typically consists of an upper-level control, which converts input EMG signal from the sensor to control command, and low-level control, for producing manipulated output (i.e., final control command) based on the force feedback from the FSR. The proportional control scheme with a force feedback mechanism was employed to control the angular position of the servomotor. The controller in the system proportionally regulates the servomotor's angular displacement via pulse width modulation (PWM) signal generated as per the reference and feedback signal [60, 61]. In the upper-level control, the root mean square (RMS) signal feature was used to generate the control signal.

An algorithm for the whole control scheme was developed on the Arduino software platform and was burned on the microcontroller unit (Arduino Nano). Two special conditions were incorporated into the control algorithm:

(1) During grasping of an object, when the difference between reference and feedback signals exceeds a threshold value, the actuators are turned off automatically. Since the servomotors are non-back-drivable, the object will remain in a grabbed position by the hand fingers. This approach will prevent physical damage to the target object during grasping and will save the battery.
(2) However, for switching on the motors again and opening the hand fingers, an EMG signal (from the user) is needed above another threshold value.

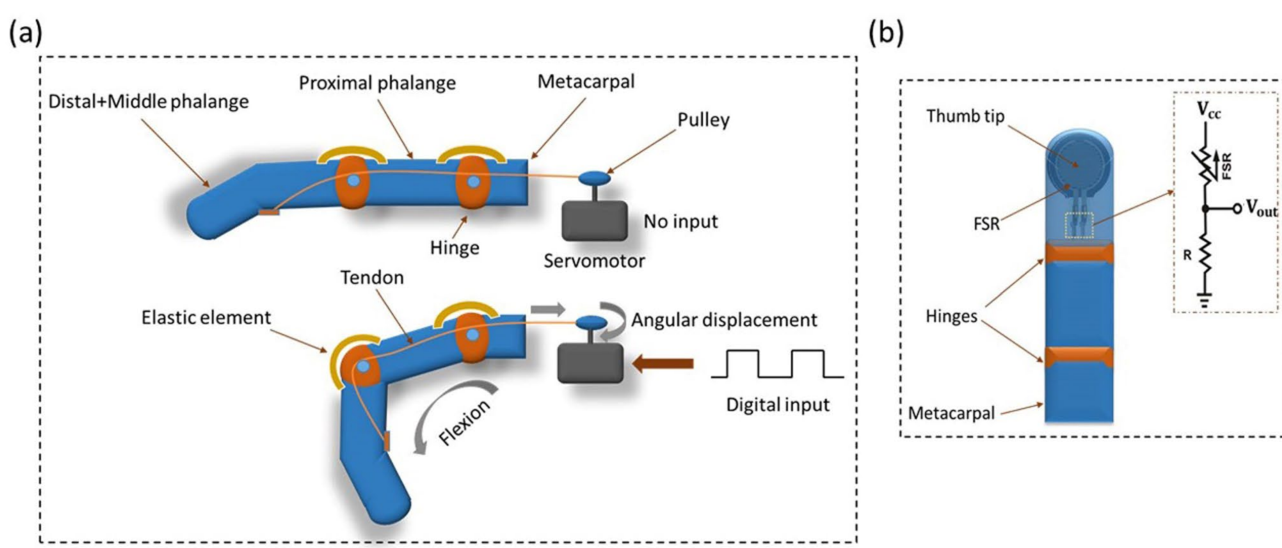


**Fig. 4** **a** Actuation scheme for hand fingers, **b** Installation of FSR at the thumb tip

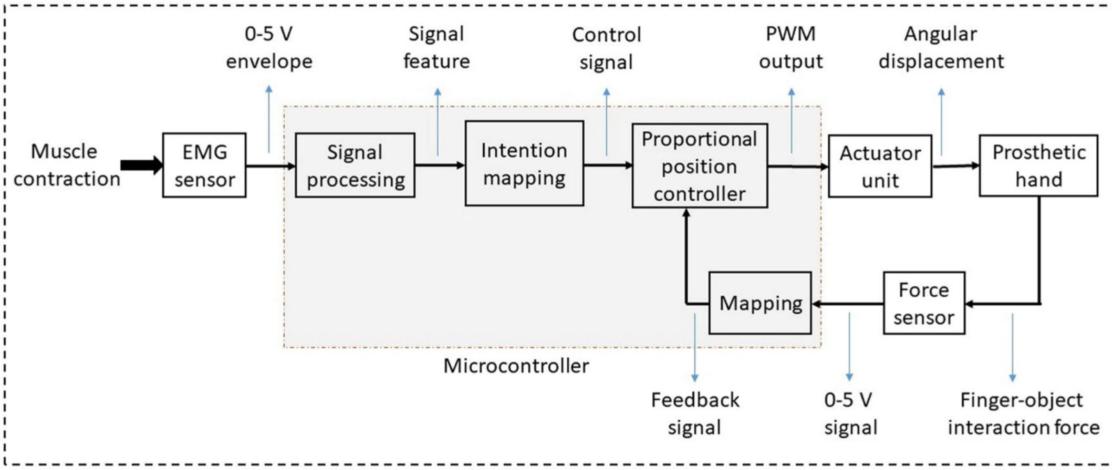


**Fig. 5** Proposed control system for the hand prosthesis

### Complete hardware setup

Figure 6a,b describes the complete setup for the 3D printed prosthetic hand. A microcontroller circuit within the installation receives analog inputs from the EMG sensor and FSR and provides a digital output (PWM) to the two servomotors. A rechargeable battery of 3.7 V, 2000 mAh with a charging module was included in the hand setup for powering the microcontroller and the motors. Silicone caps were used for all the fingers for enhancing the grasping capability of the hand. Through prosthetic socket and velcro strap, the hand was easily attachable to the residual limb of the transradial amputee.

## Hand prosthesis trial

### Subjects

The developed hand prosthesis was tested on five different subjects with transradial amputation. The details of patients who participated in the hand trial with their type and the reason for amputation are mentioned in Table 1. Before performing the experiment, written consent was taken from

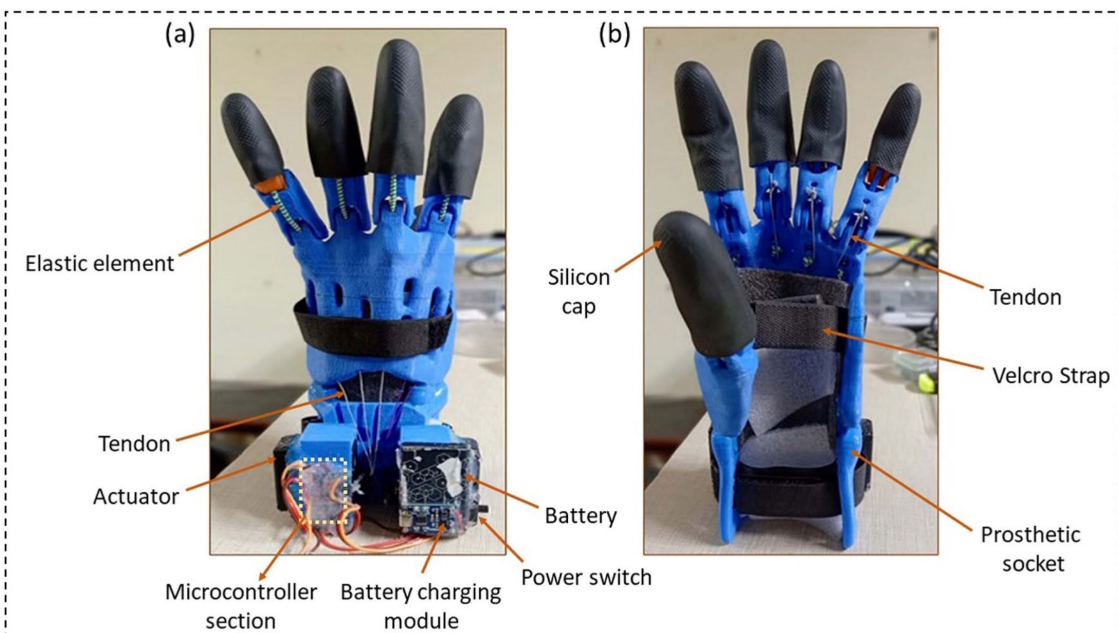


**Fig. 6** Description of the developed hand prosthesis

**Table 1** Details of amputees participated in prosthetic hand trial

| S.no | Gender | Age | Weight | Type of amputation | Reason of amputation |
|---|---|---|---|---|---|
| 1 | Male | 20 | 50 kg | Transradial(left hand) | Accident |
| 2 | Male | 50 | 85 kg | Transradial(right hand) | Accident |
| 3 | Male | 12 | 25 kg | Transradial(right hand) | Accident |
| 4 | Female | 25 | 52 kg | Transradial(right hand) | By birth |
| 5 | Male | 30 | 61 kg | Transradial(right hand) | Accident |

each subject contributed to this trial. In this work, two different hands set up were prepared for left and right-hand prosthesis user. The prosthetic hand and the EMG sensor were attached to the residual forearm stump of the individual, as shown in Fig. 7a, b. The sensor's target electrodes were positioned at flexor carpi ulnaris muscle, whereas the reference electrode was placed near the elbow region.

### Real-time testing and analysis

Subjects were instructed to contract their forearm muscles at different intensities for accomplishing activities of daily living (ADL). Participants using their intent of muscular contractions attempted several grasping tasks. Each task was tried fifteen times for 30 s duration, and the number of correct attempts was recorded. Also, the time needed for executing each grasping action was estimated for all the subjects.

To obtain the relation between the strength of muscular contraction and grasping force, the real-time value of EMG and contact force (in voltage) were measured for each grasping activity. Both the reference and feedback signals were measured considering all the subjects.

An experiment was done in which a subject was instructed to fully close/open the hand fingers frequently, and the video was captured for the same. From this recorded video, the full closing/opening time of the hand was extracted.

## Results

### Performance of designed EMG sensor

#### Sensor's raw output

The sensor's raw EMG output (i.e., amplified and filtered) and its frequency spectrum recorded for MVC of a subject are presented in Fig. 8a, b.

#### Similarity with traditional sensor

Figure 9a, b shows the EMG signal envelopes obtained using the conventional sensor and the designed sensor for four different muscular contractions of the forearm. The envelope patterns for both the sensors were observed almost similar. For the designed sensor, the EMG signal saturates at MVC however, this is not a problem as this feature was added intentionally in the sensor to obtain a

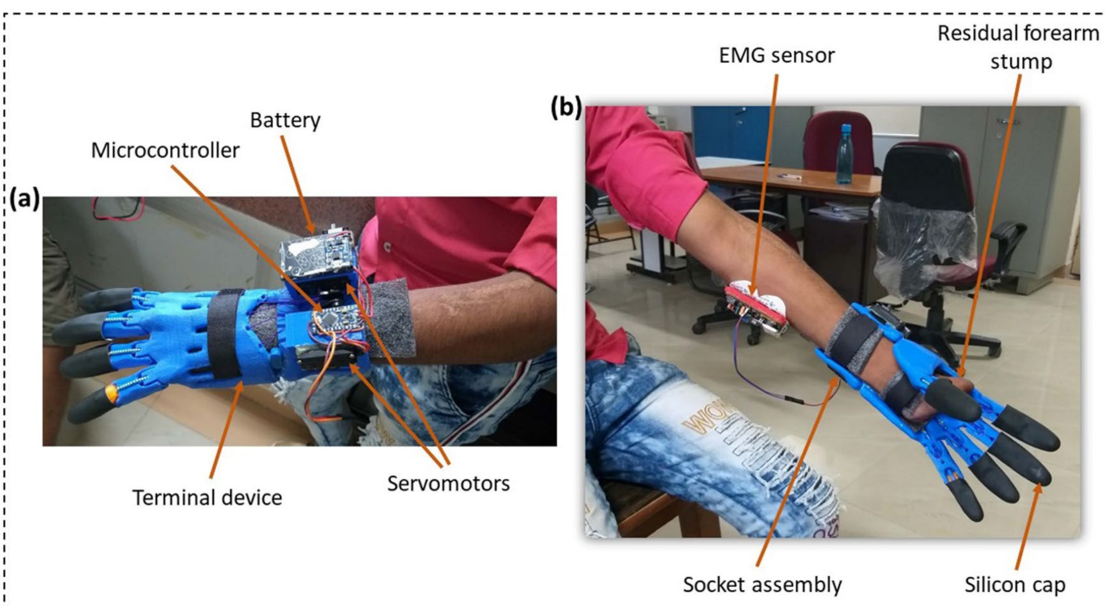


**Fig. 7** Experimental setup of hand prosthesis on a participant

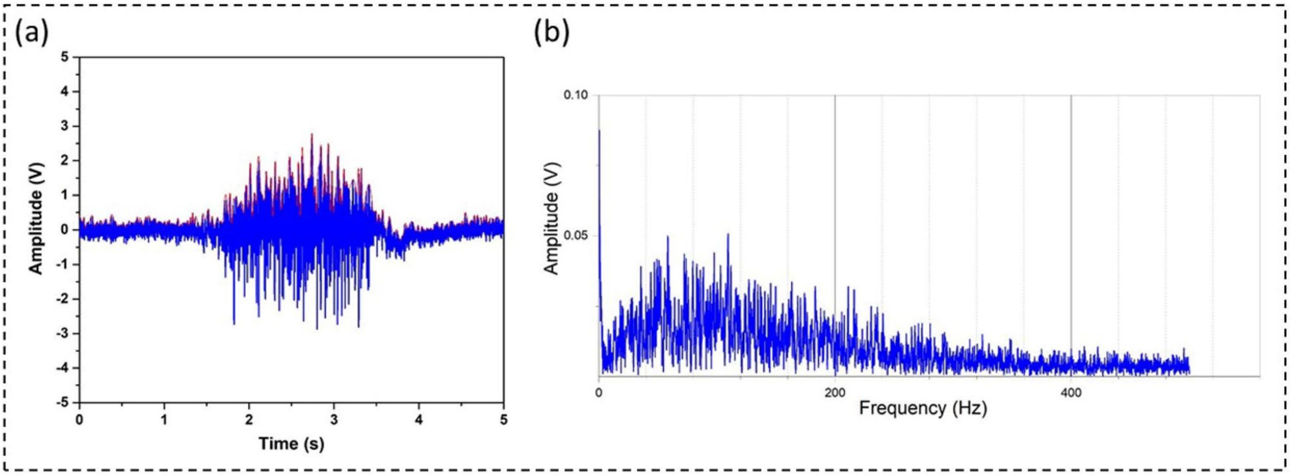


**Fig. 8** **a** Raw EMG signal from the sensor, **b** Frequency spectrum of the raw signal

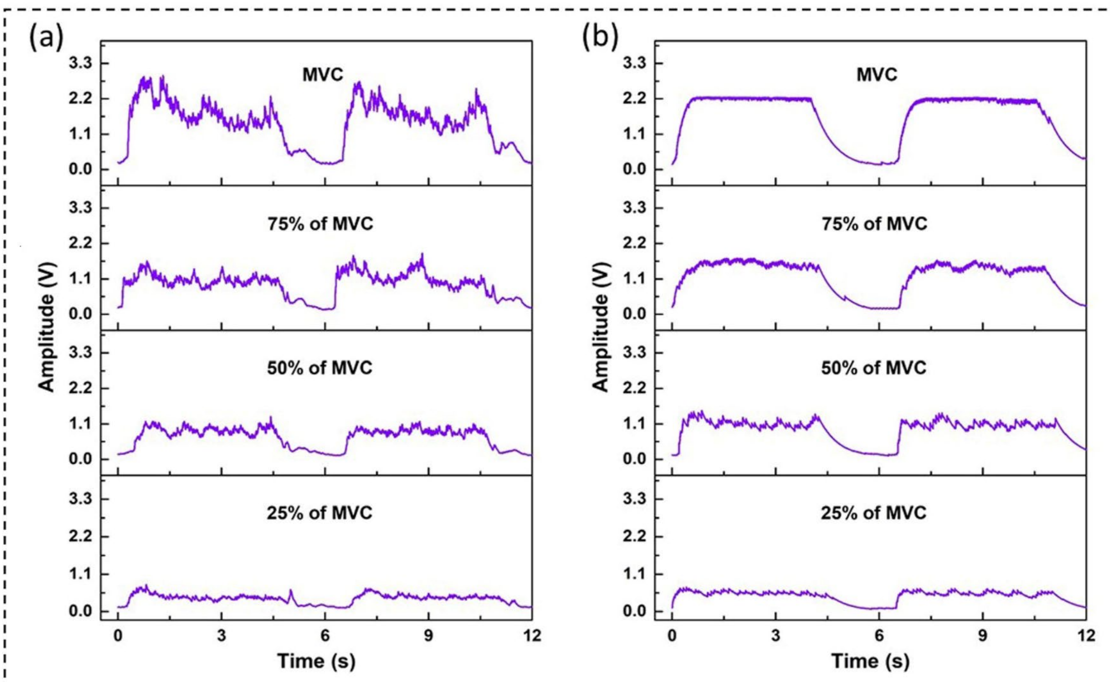


**Fig. 9** EMG envelopes obtained using. **a** Conventional sensor, b Designed sensor

smooth envelope of signal through an envelope detection scheme.

A similarity analysis performed between the signals recorded with both the sensors showed an excellent correlation coefficient ($r > 0.93$) with p-value $< 0.0001$, revealing the pairing was significantly effective.

## SNR comparison

Sensor wise SNR values determined for all the ten subjects are presented in a box whisker's plot in Fig. 10. The designed sensor displayed very high SNR values as compared to the conventional sensor.

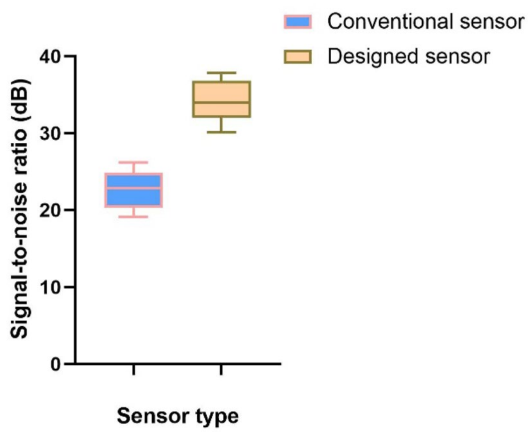


**Fig. 10** Box chart showing the variation of SNR for both the sensors

## Hand prosthesis operation

Figure 11 depicts the several activities performed by a user using a hand prosthesis, i.e., open, close, grasping sponge ball, plastic container, plastic bottle, glass container, soft drink glass, ceramic cup, and copper bottle.

The number of correct grasping activities executed by the subjects out of 15 efforts for each activity is provided in Table 2. The average percentage of success (of all the subjects) for achieving each grasping action is mentioned in the last row of the table. The overall success rate was observed at 95.4% for all the grasping efforts.

For all the subjects, the variation of time elapsed for performing each grasping action is shown in Fig. 12.

Figure 13 shows the amplitudes of EMG and contact force (i.e., grasping force) for each grasping activity. A good correlation ($r > 0.994$) was observed between both these amplitude points for all the actions. This feature revealed that the EMG signal amplitude produced by the sensor is proportional to the intensity of muscular contractions. Therefore, the grasping force of the hand is directly related to muscle contraction. Using the intent of muscular contractions, subjects can control the prehension force of the hand fingers.

The average full closing/opening time for the hand for several attempts was reported 400/550 ms. The chief reason responsible for the big difference between the closing and opening times of the prosthetic hand was the envelope detection technique used in the EMG sensor [49]. However, this can be considered as an advantage to the

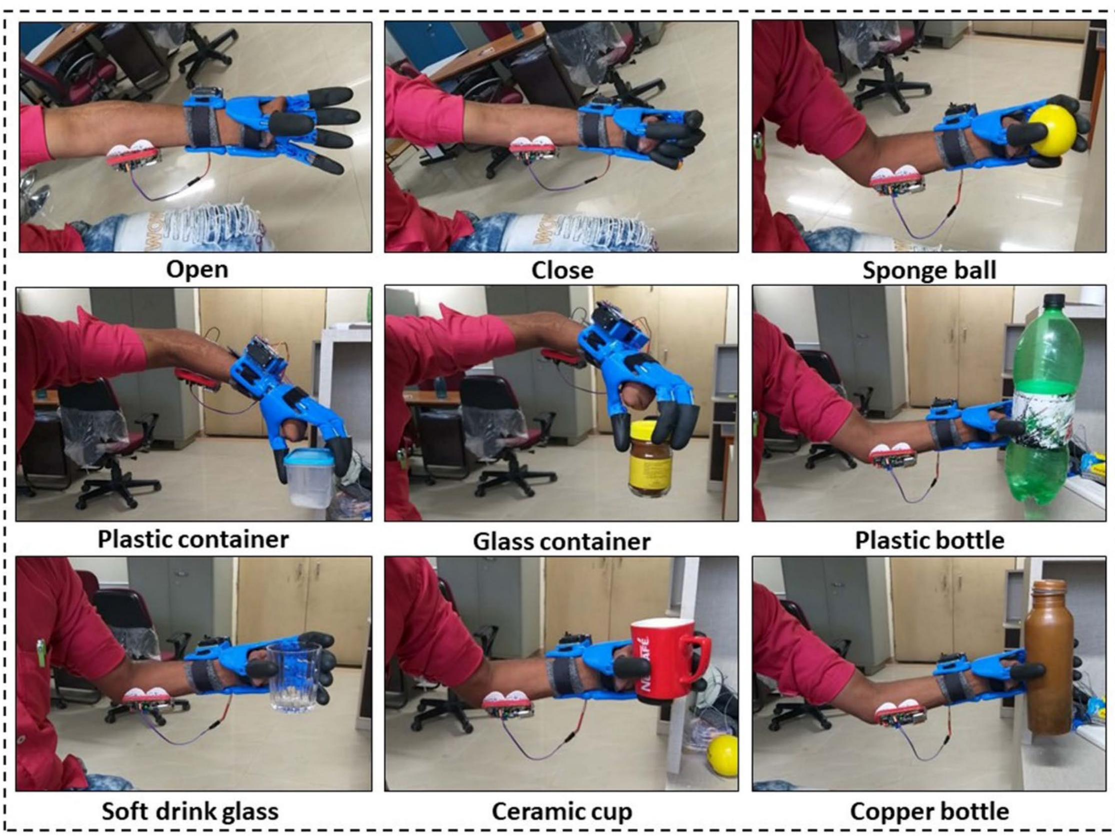


**Fig. 11** Grasping tasks performed by a prosthesis user

**Table 2** The number of correct grasping actions performed by the subjects

| Number of attempts (15) | Numbers of the correctly executed grasping task (for 30 s) | | | | | | |
|---|---|---|---|---|---|---|---|
| | Sponge all | Plastic container | Plastic bottle | Glass container | Soft drink glass (water filled) | Ceramic cup (water filled) | Copper bottle (water filled) |
| Subject 1 | 15 | 15 | 15 | 14 | 14 | 12 | 13 |
| Subject 2 | 15 | 15 | 15 | 15 | 13 | 14 | 14 |
| Subject 3 | 15 | 15 | 15 | 15 | 14 | 14 | 14 |
| Subject 4 | 15 | 15 | 14 | 14 | 14 | 13 | 15 |
| Subject 5 | 15 | 14 | 14 | 15 | 14 | 14 | 14 |
| Overall success % | 100 | 98.7 | 97.3 | 97.3 | 92 | 89.3 | 93.3 |

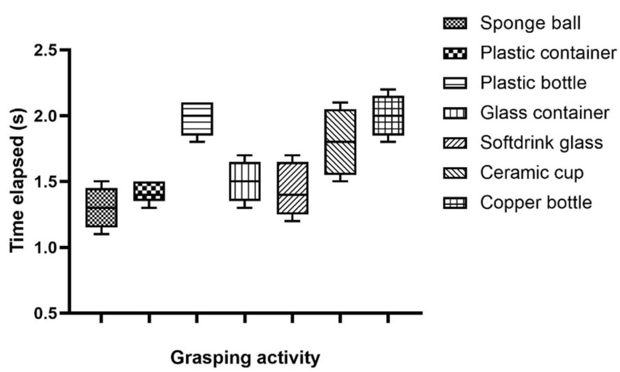


**Fig. 12** Variation of time elapsed for performing each grasping action for all the subjects

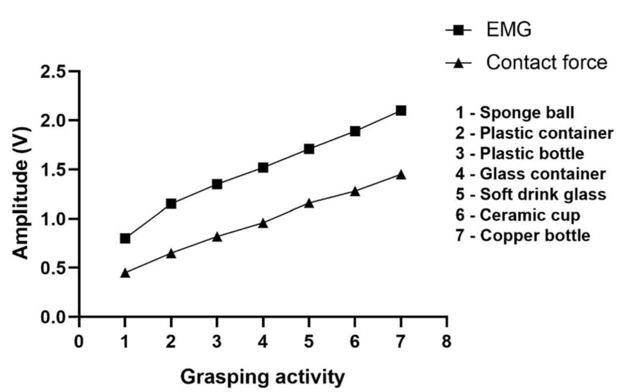


**Fig. 13** Amplitudes of EMG and contact force (in voltage) for each grasping activity

prosthetic system because a faster opening time may immediately release the grasped object.

Each participant used the hand prosthesis continuously for two hours a day, and there was no complaint of muscle fatigue. The participants revealed that the prosthesis operation was fast and usual, like a real hand. Subjects also disclosed that the prosthesis was comfortable during use.

## Discussion

The frequency spectrum of the raw EMG signal produced by the developed sensor illustrates that the signal is dominant between the frequency ranges of 20–300 Hz. The sensor was capable of generating a linear envelope proportional to the contraction intensity of the forearm muscle. For distinct hand actions, the sensor produces an EMG envelope of different magnitude and pattern.

The designed sensor showed an excellent correlation with the commercial sensor in capturing EMG signals from the forearm muscles of the subjects. Moreover, the developed sensor presented a better noise performance (i.e., SNR > 30 dB) than the conventional sensor. Under ideal conditions, the SNR of the EMG signal ranges between 10 and 50 dB [62]. These features revealed that the designed sensor could prove to be a complement to the traditional sensor for controlling prostheses. The designed sensor also has some limitations (1) it requires new disposable Ag–AgCl electrodes every time for functioning which can make the system expensive for long time use, (2) its size is too large to fit inside the socket assembly. Designing high-quality dry electrodes can completely replace the use of disposable electrodes in the sensor and can reduce the overall cost of the EMG system [63]. Moreover, using miniature-sized SMD components and professional tools for fabrication, the size of the sensor can be reduced (up to one-fourth of the original size) [49].

The underactuated 3D printed hand developed in this work was realized with the closed-loop based proportional control system in which feedback was provided by the force sensor mounted at the thumb tip. The proportional controller receiving intention based EMG input from muscular contractions provides an intuitive control output to prosthetics [13, 14]. Users with hand prosthesis attached to their remaining upper-limb were able to precisely grasp different shaped objects using the intention of muscular contractions. The FSR at the thumb tip senses contact force during finger-object interaction and avoid physical damage to fragile objects by providing feedback to the controller.

**Table 3** Comparison of the developed prosthetic hand with a commercial hand

| Parameters | Developed hand prosthesis | Ottobock sensor hand |
|---|---|---|
| Weight | 350 g (with socket) | 460 g (without socket) |
| Material | Polylactic acid (PLA) | Silicone |
| Fabrication method | 3D printing (FDM) | Moulding |
| Dimension | 175 × 85 × 35 mm | 184 × 80 × 40 mm |
| DOF | 2 | 1 |
| Number of grip pattern | 1 | 1 |
| Number of actuators | 2 | 1 |
| Actuation method | Dc motor-tendons | Dc motor-worm gear |
| Control system | Proportional | Proportional |
| Feedback | force | force |
| Full closing time | 400 ms | 300 ms |
| Battery | 3.7 V, Lithium-polymer, 2000 mAh | 7.2 V, Lithium-ion, 2200 mAh |
| EMG source | Self-designed EMG sensor | 13E200 myobock electrode |
| Price in the commercial market | Prototyping cost ($50) | $42,000 |

The inclusion of low-level control (i.e., feedback) enhances the precision in grasping up to a maximum extent [64, 65]. However, the implemented control system offers a limited number of grip patterns to the hand. Still, it is simple, low-cost, less bulky, and reliable than the pattern recognition based control system, which requires inputs from multiple sensors [26, 37, 38, 66].

A comparison of some important features between the developed hand and a commercially available hand has been presented in Table 3 [67]. The proposed prosthesis offers comparable features with the commercial hand; the only difference is the closing of the commercial hand is faster (i.e., with a closing time of 300 ms). However, the full closing time of 400–500 ms can be considered sufficiently acceptable for hand prosthesis operation [68]. Also, compared to the available multi-functional hands, the hand prosthesis provides a limited number of grip patterns; this is the main drawback of the developed system [28–31]. Increasing the number of EMG sensors, formulating a pattern recognition based control scheme, and increasing the number of actuators can enhance the number of grip patterns of the prosthetic hand. However, such an approach will also increase the overall weight of the prosthesis system and will increase complexity.

## Conclusion

In this research, a low-cost transradial prosthesis controlled by the intention of muscular contractions from the user was developed. A specific, open-source EMG sensor was designed for reliable detection of the muscular activity from the residual upper-limb of persons with amputation. The sensor performance was validated with a traditional EMG sensor. An underactuated hand prosthesis was developed with a closed-loop based proportional control system that utilizes EMG input (from the sensor) as a reference and grip force input (from the FSR) as feedback.

Further, the developed prosthetic hand was tested on five subjects with below-elbow amputation for the execution of various grasping tasks (with a success rate > 95%). The intended control system allowed faster and intuitive operation of hand with control on grip force.

Although the developed hand has limited functionality, it is still simple, low-cost, lightweight, have fast operating speed, delivers control like a natural hand, and does not require prior training sessions. These features are sufficient to increase the acceptability rate of hand prostheses among amputees. Even though the proposed prosthesis system appears to be less clinically acceptable as compared to the commercial one, with enhanced features and design, it can make a real clinical difference to be used by the user.

**Acknowledgements** The authors would like to thank the Science and Engineering Research Board (SERB), Department of Science and Technology, Government of India, for funding this project.

**Funding** Science and Engineering Research Board (SERB) [Grant No. CRG/2018/001612], Department of Science and Technology, Government of India.

## Compliance with ethical standards

**Conflict of interest** The authors declare that there are no conflicts of interest.

**Ethical approval (involvement of animals)** This article does not contain any studies with animals performed by any of the authors.

**Ethical approval (involvement of human subjects)** This article involves surface EMG data acquisition from various human subjects. Ethical approval was taken from the Ethical committee, Institute of medical sciences, BHU, Varanasi before performing this experiment. All procedures performed in studies involving human participants were in accordance with the ethical standards of the institutional and/or national research committee and with the 1964 Helsinki declaration and its later amendments or comparable ethical standards.

**Informed consent** Informed consent was obtained from all individual participants involved in the study. In addition, informed consent was taken from the parents of minor participants for this study.

## References


1. Slade P, Akhtar A, Nguyen M, Bretl T (2015) Tact: design and performance of an open-source, affordable, myoelectric prosthetic hand. In: 2015 IEEE International Conference on Robotics and Automation (ICRA). IEEE, Seattle, WA, USA, pp 6451–6456
2. World Report on Disability. 350
3. Sahu A, Sagar R, Sarkar S, Sagar S (2016) Psychological effects of amputation: A review of studies from India. Ind Psychiatry J 25:4–10. https://doi.org/10.4103/0972-6748.196041
4. Geethanjali P (2016) Myoelectric control of prosthetic hands: state-of-the-art review. Med Devices (Auckl) 9:247–255. https://doi.org/10.2147/MDER.S91102
5. Uellendahl J (2017) Myoelectric versus body-powered upper-limb prostheses: a clinical perspective. JPO 29:25. https://doi.org/10.1097/JPO.0000000000000151
6. Cordella F, Ciancio AL, Sacchetti R, Davalli A, Cutti AG, Guglielmelli E, Zollo L (2016) Literature review on needs of upper limb prosthesis users. Front Neurosci. https://doi.org/10.3389/fnins.2016.00209
7. Prakash A, Sahi AK, Sharma N, Sharma S (2020) Force myography controlled multifunctional hand prosthesis for upper-limb amputees. Biomed Signal Process Control 62:102122. https://doi.org/10.1016/j.bspc.2020.102122
8. Cheesborough JE, Smith LH, Kuiken TA, Dumanian GA (2015) Targeted muscle reinnervation and advanced prosthetic arms. Semin Plast Surg 29:62–72. https://doi.org/10.1055/s-0035-1544166
9. Fifer MS, Acharya S, Benz HL, Mollazadeh M, Crone NE, Thakor NV (2012) towards electrocorticographic control of a dexterous upper limb prosthesis. IEEE Pulse 3:38–42. https://doi.org/10.1109/MPUL.2011.2175636
10. AL-Quraishi MS, Elamvazuthi I, Daud SA, Parasuraman S, Borboni A (2018) EEG-based control for upper and lower limb exoskeletons and prostheses: a systematic review. Sensors (Basel). https://doi.org/10.3390/s18103342
11. Asghari Oskoei M, Hu H (2007) Myoelectric control systems: a survey. Biomed Signal Process Control 2:275–294. https://doi.org/10.1016/j.bspc.2007.07.009
12. Parker P, Englehart K, Hudgins B (2006) Myoelectric signal processing for control of powered limb prostheses. J Electromyogr Kinesiol 16:541–548. https://doi.org/10.1016/j.jelekin.2006.08.006
13. George JA, Davis TS, Brinton MR, Clark GA (2020) Intuitive neuromyoelectric control of a dexterous bionic arm using a modified Kalman filter. J Neurosci Methods 330:108462. https://doi.org/10.1016/j.jneumeth.2019.108462
14. Jiang N, Rehbaum H, Vujaklija I, Graimann B, Farina D (2014) Intuitive, online, simultaneous, and proportional myoelectric control over two degrees-of-freedom in upper limb amputees. IEEE Trans Neural Syst Rehabil Eng 22:501–510. https://doi.org/10.1109/TNSRE.2013.2278411
15. Milosevic B, Benatti S, Farella E (2017) Design challenges for wearable EMG applications. In: Design, Automation Test in Europe Conference Exhibition (DATE), pp 1432–1437
16. Liu J, Zhou P (2013) A novel myoelectric pattern recognition strategy for hand function restoration after incomplete cervical spinal cord injury. IEEE Trans Neural Syst Rehabil Eng 21:96–103. https://doi.org/10.1109/TNSRE.2012.2218832
17. Tavakoli M, Benussi C, Lourenco JL (2017) Single channel surface EMG control of advanced prosthetic hands. Expert Syst Appl 79:322–332. https://doi.org/10.1016/j.eswa.2017.03.012
18. Pancholi S, Joshi AM (2018) Portable EMG data acquisition module for upper limb prosthesis application. IEEE Sens J 18:3436–3443. https://doi.org/10.1109/JSEN.2018.2809458
19. Muzumdar A (2004) Powered upper limb prostheses. Springer, Berlin
20. Prakash A, Sharma S (2020) A low-cost system to control prehension force of a custom-made myoelectric hand prosthesis. Res Biomed Eng 36:237–247. https://doi.org/10.1007/s42600-020-00064-w
21. Cranny A, Cotton DPJ, Chappell PH, Beeby SP, White NM (2005) Thick-film force, slip and temperature sensors for a prosthetic hand. Meas Sci Technol 16:931–941. https://doi.org/10.1088/0957-0233/16/4/005
22. Engeberg ED, Meek SG (2013) adaptive sliding mode control for prosthetic hands to simultaneously prevent slip and minimize deformation of grasped objects. IEEE/ASME Trans Mechatron 18:376–385. https://doi.org/10.1109/TMECH.2011.2179061
23. Borisov II, Borisova OV, Krivosheev SV, Oleynik RV, Reznikov SS Prototyping of EMG-controlled prosthetic hand with sensory system. 5
24. Cipriani C, Controzzi M, Carrozza MC (2011) The SmartHand transradial prosthesis. J NeuroEng Rehabil 8:29. https://doi.org/10.1186/1743-0003-8-29
25. Furui A, Eto S, Nakagaki K, Shimada K, Nakamura G, Masuda A, Chin T, Tsuji T (2019) A myoelectric prosthetic hand with muscle synergy–based motion determination and impedance model–based biomimetic control. Sci Robot. https://doi.org/10.1126/scirobotics.aaw6339
26. Wang N, Lao K, Zhang X (2017) Design and myoelectric control of an anthropomorphic prosthetic hand. J Bionic Eng 14:47–59. https://doi.org/10.1016/S1672-6529(16)60377-3
27. Williams MR, Walter W (2015) Development of a prototype over-actuated biomimetic prosthetic hand. PLoS ONE 10:e0118817. https://doi.org/10.1371/journal.pone.0118817
28. bebionic hand. https://www.ottobockus.com/prosthetics/upper-limb-prosthetics/solution-overview/bebionic-hand/. Accessed 21 Oct 2019
29. i-limb quantum | Touch Bionics. http://touchbionics.com/products/active-prostheses/i-limb-quantum. Accessed 12 Apr 2019
30. Michelangelo prosthetic hand. https://www.ottobockus.com/prosthetics/upper-limb-prosthetics/solution-overview/michelangelo-prosthetic-hand/. Accessed 12 Apr 2019
31. VINCENTevolution 3. https://vincentsystems.de/en/prosthetics/vincent-evolution-3/. Accessed 12 Apr 2019
32. Atzori M, Müller H (2015) Control capabilities of myoelectric robotic prostheses by hand amputees: a scientific research and market overview. Front Syst Neurosci 9:162. https://doi.org/10.3389/fnsys.2015.00162
33. Li G, Schultz AE, Kuiken TA (2010) Quantifying pattern recognition—based myoelectric control of multifunctional transradial prostheses. IEEE Trans Neural Syst Rehabil Eng 18:185–192. https://doi.org/10.1109/TNSRE.2009.2039619
34. Daley H, Englehart K, Hargrove L, Kuruganti U (2012) High density electromyography data of normally limbed and transradial amputee subjects for multifunction prosthetic control. J

Electromyogr Kinesiol 22:478–484. https://doi.org/10.1016/j.jelekin.2011.12.012
35. Hamner SR, Narayan VG, Donaldson KM (2013) Designing for scale: development of the remotion knee for global emerging markets. Ann Biomed Eng 41:1851–1859. https://doi.org/10.1007/s10439-013-0792-8
36. Prakash A, Sharma S (2020) Development of an affordable myoelectric hand for transradial amputees: international journal of biomedical and clinical. Engineering 9:1–15. https://doi.org/10.4018/IJBCE.2020010101
37. Al-Timemy AH, Bugmann G, Escudero J, Outram N (2013) Classification of finger movements for the dexterous hand prosthesis control with surface electromyography. IEEE J Biomed Health Inform 17:608–618. https://doi.org/10.1109/JBHI.2013.2249590
38. Khushaba RN, Al-Timemy AH, Al-Ani A, Al-Jumaily A (2017) A framework of temporal-spatial descriptors-based feature extraction for improved myoelectric pattern recognition. IEEE Trans Neural Syst Rehabil Eng 25:1821–1831. https://doi.org/10.1109/TNSRE.2017.2687520
39. Chandrapal M (2012) Intelligent Assistive Knee Orthotic Device Utilizing Pneumatic Artificial Muscles
40. Chatfield LT, Fortune BC, McKenzie LR, Pretty CG (2018) Implementation of a particle filter to estimate torque from electromyography. IFAC-PapersOnLine 51:327–332. https://doi.org/10.1016/j.ifacol.2018.11.620
41. Chatfield LT, Fortune BC, McKenzie LR, Pretty CG (2019) Development of an assist-as-need controller for an upper-limb exoskeleton with voluntary torque estimate. american society of mechanical engineers digital collection
42. Sankai Y (2011) HAL: hybrid assistive limb based on cybernics. In: Kaneko M, Nakamura Y (eds) Robotics research. Springer, Berlin, Heidelberg, pp 25–34
43. Fortune BC, Pretty CG, Chatfield LT, McKenzie LR, Hayes MP (2019) Low-cost active electromyography. HardwareX 6:e00085. https://doi.org/10.1016/j.ohx.2019.e00085
44. Prakash A, Kumari B, Sharma S (2019) A low-cost, wearable sEMG sensor for upper limb prosthetic application. J Med Eng Technol 43:235–247. https://doi.org/10.1080/03091902.2019.1653391
45. De Luca CJ (1997) The use of surface electromyography in biomechanics. J Appl Biomech 13:135–163. https://doi.org/10.1123/jab.13.2.135
46. Gerdle B, Karlsson S, Day S, Djupsjöbacka M (1999) Acquisition, processing and analysis of the surface electromyogram. In: Windhorst U, Johansson H (eds) Modern techniques in neuroscience research. Springer, Berlin, pp 705–755
47. Sharma N, Prakash A, Sahi AK, Sharma N, Sharma S (2020) Multimodal sensor to measure the concurrent electrical and mechanical activity of muscles for controlling a hand prosthesis. Instrum Sci Technol. https://doi.org/10.1080/10739149.2020.1804932
48. Wang J, Tang L, Bronlund E (2013) Surface EMG signal amplification and filtering. Int J Comput Appl 82:15–22. https://doi.org/10.5120/14079-2073
49. Prakash A, Sharma S, Sharma N (2019) A compact-sized surface EMG sensor for myoelectric hand prosthesis. Biomed Eng Lett 9:467–479. https://doi.org/10.1007/s13534-019-00130-y
50. Farina D (2006) Interpretation of the surface electromyogram in dynamic contractions. Exerc Sport Sci Rev 34:121–127
51. Heywood S, Pua YH, McClelland J, Geigle P, Rahmann A, Bower K, Clark R (2018) Low-cost electromyography – Validation against a commercial system using both manual and automated activation timing thresholds. J Electromyogr Kinesiol 42:74–80. https://doi.org/10.1016/j.jelekin.2018.05.010
52. Tatarian K, Couceiro MS, Ribeiro EP, Faria DR (2018) Stepping-stones to Transhumanism: an EMG-controlled low-cost prosthetic hand for academia. In: 2018 International Conference on Intelligent Systems (IS). IEEE, Funchal-Madeira, Portugal, pp 807–812
53. Lobo-Prat J, Kooren PN, Stienen AH, Herder JL, Koopman BF, Veltink PH (2014) Non-invasive control interfaces forintention detection in active movement-assistive devices. J NeuroEng Rehabil 11:168. https://doi.org/10.1186/1743-0003-11-168
54. Agostini V, Knaflitz M (2012) An algorithm for the estimation of the signal-to-noise ratio in surface myoelectric signals generated during cyclic movements. IEEE Trans Biomed Eng 59:219–225. https://doi.org/10.1109/TBME.2011.2170687
55. Enabling The Future. In: Enabling The Future. https://enablingthefuture.org/. Accessed 20 Oct 2020
56. Melocchi A, Parietti F, Loreti G, Maroni A, Gazzaniga A, Zema L (2015) 3D printing by fused deposition modeling (FDM) of a swellable/erodible capsular device for oral pulsatile release of drugs. J Drug Deliv Sci Technol 30:360–367. https://doi.org/10.1016/j.jddst.2015.07.016
57. Pylatiuk C, Schulz S, Kargov A, Bretthauer G (2004) Two multiarticulated hydraulic hand prostheses. Artif Organs 28:980–986. https://doi.org/10.1111/j.1525-1594.2004.00014.x
58. Prakash A, Sharma N, Sharma S (2020) Novel force myography sensor to measure muscle contractions for controlling hand prostheses. Instrum Sci Technol 48:43–62. https://doi.org/10.1080/10739149.2019.1655441
59. Schoepp KR, Dawson MR, Schofield JS, Carey JP, Hebert JS (2018) Design and integration of an inexpensive wearable mechanotactile feedback system for myoelectric prostheses. IEEE J Transl Eng Health Med 6:1–11. https://doi.org/10.1109/JTEHM.2018.2866105
60. Fougner A, Stavdahl O, Kyberd PJ, Losier YG, Parker PA (2012) Control of upper limb prostheses: terminology and proportional myoelectric control-a review. IEEE Trans Neural Syst Rehabil Eng 20:663–677. https://doi.org/10.1109/TNSRE.2012.2196711
61. Lenzi T, De Rossi SMM, Vitiello N, Carrozza MC (2011) Proportional EMG control for upper-limb powered exoskeletons. In: 2011 Annual International Conference of the IEEE Engineering in Medicine and Biology Society. IEEE, Boston, MA, pp 628–631
62. Sinderby C, Lindström L, Grassino AE (1995) Automatic assessment of electromyogram quality. J Appl Physiol 79:1803–1815. https://doi.org/10.1152/jappl.1995.79.5.1803
63. Jamal MZ, Kim K-S (2018) A finely machined toothed silver electrode surface for improved acquisition of EMG signals. In: 2018 IEEE Sensors Applications Symposium (SAS). IEEE, Seoul, pp 1–5
64. Sebastian F, Fu Q, Santello M, Polygerinos P (2017) Soft robotic haptic interface with variable stiffness for rehabilitation of neurologically impaired hand function. Front Robot AI. https://doi.org/10.3389/frobt.2017.00069
65. Zhu G, Duan X, Deng H (2013) Hybrid Force-Position Fuzzy Control for a Prosthetic Hand. In: Lee J, Lee MC, Liu H, Ryu J-H (eds) Intelligent Robotics and Applications. Springer, Berlin , pp 415–426
66. Geng W, Du Y, Jin W, Wei W, Hu Y, Li J (2016) Gesture recognition by instantaneous surface EMG images. Sci Rep 6:36571. https://doi.org/10.1038/srep36571
67. Myoelectric Speed hands. https://www.ottobockus.com/prosthetics/upper-limb-prosthetics/solution-overview/myoelectric-devices-speedhands/. Accessed 12 Apr 2019
68. Chadwell A, Kenney L, Thies S, Galpin A, Head J (2016) The reality of myoelectric prostheses: understanding what makes these devices difficult for some users to control. Front Neurorobot. https://doi.org/10.3389/fnbot.2016.00007