Research on Biomedical Engineering (2020) 36:237–247
https://doi.org/10.1007/s42600-020-00064-w

ORIGINAL ARTICLE

A low-cost system to control prehension force of a custom-made myoelectric hand prosthesis

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Received: 22 October 2019 / Accepted: 22 April 2020 / Published online: 6 June 2020
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Abstract

Purpose In order to achieve stable and dexterous grasping of objects, prehension force control is quite a significant parameter for prosthetic hands. Commercially available hands such as bebionic, i-limb quantum and Michelangelo offer the precise grasping capability to perform activities of daily living (ADLs). However, the cost of such hands is too expensive for amputees residing in low-income countries.

Methods This paper introduces a low-cost, simple and efficient system for controlling the prehension force of a self-designed myoelectric prosthetic hand. A hand prototype was developed employing 3D printing technology and an intrinsic actuation approach. The hand fingers were equipped with a pre-calibrated force sensor for the online estimation of the grasp force. A closed-loop proportional-derivative (PD) based position control system was designed considering actuator as plant, electromyography (EMG) as a reference and grasp force as a feedback signal.

Results The results showed highly improved parameters, i.e. overshoot, offset and settling time of the proposed system than a simple open-loop system. These parameters guarantee faster closing of hand fingers and the production of accurate prehension force during finger-object interaction.

Conclusion Further, the myoelectric hand with a developed control scheme was successfully tested on five different transradial amputees for performing precise and faster grasping of different shaped objects.

Keywords Electromyography · Myoelectric prosthesis · Prehension force · Force sensor · Position control

Introduction

The technology involved in the design and development of myoelectric prosthesis has shown rapid progress in restoring the lost capabilities of upper limb amputees by utilizing the sEMG signal from their residual limb. A myoelectric prosthesis typically consists of a device for detecting EMG signal, a controller that converts these signals to control commands for driving actuators coupled with the prosthetic device, by employing real-time learning (Asghari Oskoei and Hu 2007; Parker et al. 2006; Prakash et al. 2019b).

Bebionic v3, i-limb quantum, Ottobock and Utah arm are some of the commercially available myoelectric hands that can offer several features like less weight, higher degrees of freedom (DOF), excellent grasping capability and considerable grip strength, but their cost is quite expensive (Bebionic hand 2019; i-limb quantum 2019; Myoelectric Speed hands 2019; Motion control, Utah arm 2019). However, the simple hand prosthesis with one or two DOF is incapable of providing precision in grasping and dexterity (Sono and Menegaldo 2009). These are the main reasons responsible for the rejection of hand prosthesis by amputees. While designing prosthetic hand, a trade-off between cost and functionality should be done such that the device becomes affordable but still maintaining the performance equivalent to the commercial hands (Borisov et al. 2017; Slade et al. 2015; Wang et al. 2017).

The control system in present prosthetic hands typically comprises an upper-level control for mapping input EMG signal to control command and a low-level control for producing final output based on position, velocity or force feedback (Muzumdar 2004). The feedback is provided by the various tactile sensors embedded within the prosthetics. The low-level control is quite essential because the current prosthesis suffers

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from time delays and also lacks proprioceptive feedback to subjects (Engeberg et al. 2008).

A control strategy or upper-level control in myoelectric prosthesis translates the extracted features of the EMG signal to control command for prosthetic device actuation. Pattern recognition-based myoelectric control requires classification procedure but provides more functions to prosthetic hand as compared to the non-pattern recognition based control scheme (Li et al. 2010). Proportional control is a non-pattern recognition-based control scheme that provides natural control and faster-grasping capability to the hand (Fougner et al. 2012; Lenzi et al. 2011). Such a control scheme relies on factors like characteristics of EMG detection unit, position of electrode on the skin and muscle fatigue for proper generation of control commands (Herle et al. 2012; Hudgins et al. 1993).

Low-level force control is highly recommended for achieving accurate and delicate grasping of the hand prosthesis (Cranny et al. 2005; Engeberg and Meek 2013). The tactile sensors employed for force control must be robust, low-power, low cost and small such that it can be easily installed at the fingertip for monitoring force applied on the sensor (Chappell and Elliott 2003). In the closed-loop operation of the prosthetic hand, the system senses the contact force of object-finger interaction and compares it to the desired force level from EMG signal for generating the effective control signal for driving actuators (Sono and Menegaldo 2009). Qiushi Fu et al. proposed a hybrid controller based on mechanotactile haptic feedback for improving fine control of grasping force during object pick-and-place tasks with prosthetic hand (Sebastian et al. 2017). A fuzzy-based PID controller was designed for accurate force control of multifunctional prosthetic hand incorporating force-sensitive resistor (FSR) (Ghazali et al. 2017). Gaoke Zhu et al. (2013) presented an adaptive fuzzy-based PID control strategy for effective force control of a prosthetic hand. Object slippage detection was done using vibrations persuaded in force signals captured by sensors installed in the fingers of the prosthetic hand (Schwarz 2016). Loredana Zollo et al. (2019) presented a closed-loop system for controlling real-time force and slippage control of bionic hands employing tactile sensations via neural interfaces. During the closed-loop operation, the tactile sensors integrated into the fingers were utilized to provide force and slippage information for the amputee.

Despite several advancements in sensor technology, control system and design, hand prosthesis with sufficient functionality and suitable dexterity are unavailable to the majority of amputees residing in developing countries. The chief reasons are the overall cost of the hand, its complexity, controllability, reliability and lack of resources (Geethanjali 2016; Godfrey et al. 2018; Prakash and Sharma 2020; Wang et al. 2017).

This paper proposes a low-cost, effortless and effective scheme for controlling the grasp force of a custom-made myoelectric hand. In this work, a 3D-printed hand prototype was developed and actuated using an intrinsic actuation approach. A highly sensitive and calibrated tactile sensor made from force-sensitive resistor (FSR) was installed at the fingertip for measurement of grasp force. A closed-loop proportional-derivative (PD)-based position control system was designed for controlling the prehension force of hand fingers. The desired flexion of fingers was provided by the EMG signal from the sensor, whereas the actual force applied by the fingers on the object was estimated using the tactile sensor installed at the fingertip. The transfer function of the plant model (consisting of actuators) was determined using a system identification technique. The designed control scheme with feedback showed better performance parameters as compared to the open-loop control. Further, the myoelectric hand with the developed control scheme was successfully tested on amputees for performing dexterous grasping operations.

Methods

Prosthetic hand design

3D printing

A 3D-printed hand was proposed in this work to reduce the cost, weight and enhance the cosmetic appeal. 3D printing technology permits us to construct lighter, proper strength and customized objects as per our design. With a low mass, the 3D-printed hand could potentially reduce the pain and fatigue caused by mass distribution over softer tissues (Ten Kate et al. 2017). A five-finger hand model from E-NABLE was custom-made and printed using PLA (polylactic acid) filaments of 1.75 mm diameter with 40% filling density and extruder temperature set at 230 °C (e-NABLE phoenix hand v2 2019). The hand parts were printed in 20 h using a dimension-raised 3D printer and were assembled and redesigned, as shown in Fig. 1b. The final dimension of the obtained 3D-printed hand was 175 mm × 85 mm × 35 mm. The hand fingers were equipped with silicon fingertip for enhancing its grasping capability.

Actuation

The hand fingers were intrinsically actuated by two servomotors (i.e. motors were located inside the palm). Such a scheme promotes a more practical design for dexterous operation of hand (Kargov et al. 2004). The flexion of fingers was provided by motor-driven tendons, whereas their extension was done using the elastic bands attached between the finger joints. The thumb was driven by one of the servomotors, while the other servomotor operated the rest four fingers. High-torque, low-weight metal-geared servomotor (MG-996R) was used here as

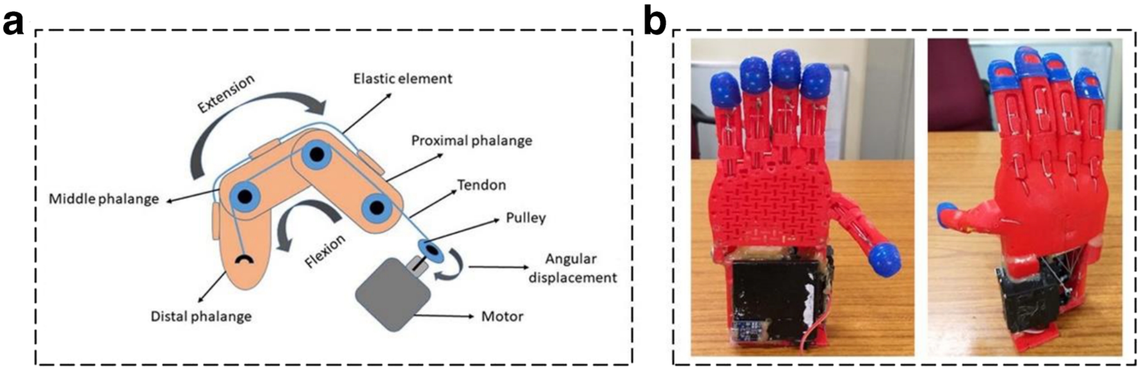


Fig. 1 **a** Finger actuation scheme, **b** developed 3D-printed hand prototype

the finger actuator providing maximum torque up to 13 kgf cm at an operating voltage of 5 V. Pulleys attached to the motors were used to manipulate the tendons to provide flexion of fingers. A high-tension fishing line of 0.66 mm diameter was employed as tendons. Figure 1a shows the actuation scheme for the fingers. Since the fingers had three joints, the developed hand was capable of producing a total of three degrees of freedom (DOF) with only two actuators. This type of underactuated mechanism offers the grasping of objects in an entirely natural way as analogues to human hand (Deimel and Brock 2016). Arduino nano microcontroller was installed inside the hand assembly for generating a pulse width modulation (PWM) output signal to drive the servomotors using EMG input. Figure 1b shows the fully actuated 3D-printed hand prototype.

EMG sensor

A surface EMG sensing module was fabricated for producing reference input to control the operation of the prosthetic hand. The sensing module typically consists of the electrode interface, the signal conditioning circuitry and the power supply unit, all embedded in a single structure. The various stages involved in the signal conditioning unit of the sensor are pre-amplifier, band-pass filter, post-amplifier and envelope detector (Prakash and Kumari 2019). The sensor provides EMG output as a linear envelope of 0–5 V proportional to the muscular contractile force. For acquiring EMG signal, the target electrodes of the sensor were placed at flexor carpi radialis and flexor carpi ulnaris muscles while the reference electrode at elbow region as these muscles in the forearm are responsible for the flexion of fingers and wrist movement (Hermens et al. 1999). Figure 2 shows the attachment sensor on the residual forearm stump of amputees for acquiring their EMG signal. The EMG output was applied to the analog input port of the microcontroller for the generation of the control command (i.e. PWM signals) to drive actuators.

Upper-level control

The upper-level control is mainly responsible for the generation of a control command from the captured sEMG signal to drive prosthetics. The control strategy determines how flexion of prosthetic hand fingers will be performed using the muscular contractions (Muzumdar 2004). The desired closing of fingers produces a force to grasp the objects. The operation of digits can be threshold-based or proportional. In threshold control, if the EMG signal level is higher than a particular value, the command will be generated to close the fingers; otherwise, the hand will be in an open state. This control scheme provides either fully open or fully close operation of the hand with fixed prehension force, making the control quite unnatural (Asghari Oskoei and Hu 2007; Herle et al. 2012). On the other side, the proportional scheme provides the flexion of fingers as per the intensity of the captured EMG signal. Here the prehension force of the fingers is proportional to the strength of EMG signal (Fougner et al. 2012). Up to some extent, this control scheme provides an intuitive operation to

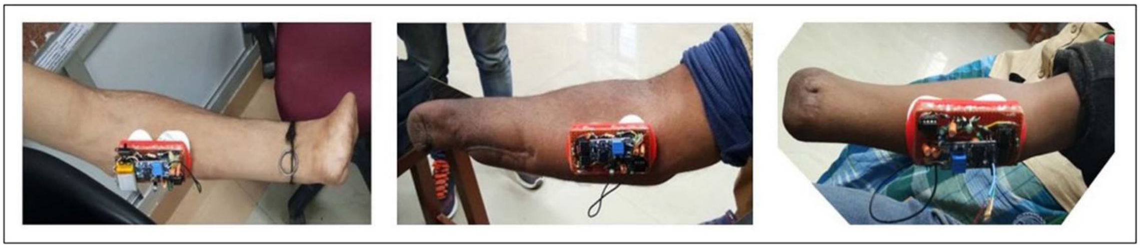


Fig. 2 EMG sensor attachment on residual forearm stump of amputees

prosthetic hand with control on its grasp force. However, to achieve accurate grasping of objects (with slippage free operation), online measurement of prehension force is a must to provide feedback to the controller (Cranny et al. 2005; Engeberg and Meek 2013). In that case, the open-loop based control system is not sufficient for smooth and accurate operation of the prosthesis. This work deals with the design of a closed-loop control system that combines the concept of upper-level and low-level control for the myoelectric prosthesis.

Tactile sensor

A piezoresistive-based tactile sensor was designed using FSR to estimate the contact force during finger-object interaction. The FSR sensing portion was encased in a polydimethylsiloxane (PDMS) structure shown in Fig. 3a for the proper distribution of grip force over the contact surface area. An appropriate voltage divider circuit is shown in Fig. 3b was employed for converting the change in resistance of FSR to the voltage output (Prakash et al. 2019a). The measured output of the sensor is a 0–5 V linear envelope similar to that of the EMG sensor. Finally, the sensor was fixed to the fingertip using a silicon cap, as shown in Fig. 3c. The sensor at the fingertip serves as a feedback element for providing contact force information during object grasping. FSR, as a force measuring element, offers various advantages like low-cost, small size, lightweight, robustness, and good accuracy. Also, FSR is an optimal sensor in the prosthetic application, which provides a reliable measurement of force over the existing state of sensor technologies (Schoepp et al. 2018). The sensor output was applied at the analog input port of the microcontroller for getting an estimated value of prehension force during finger-object interaction.

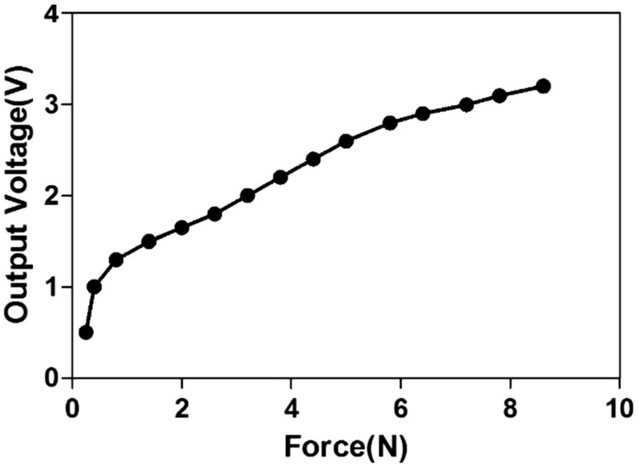


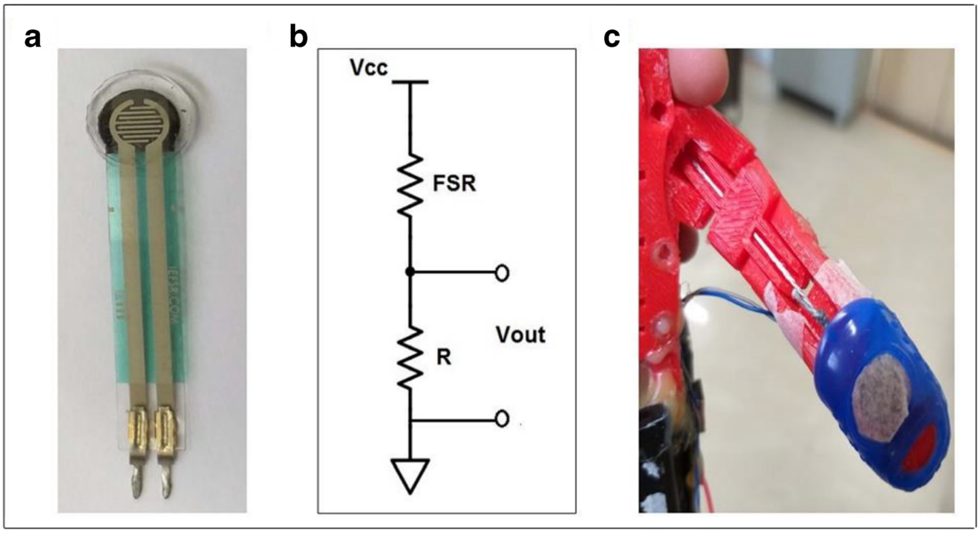
Fig. 4 Force-voltage calibration curve for the FSR

A force-voltage calibration curve shown in Fig. 4 was obtained for determining the contact force (in newton) exerted on the FSR installed at the fingertip. To perform this calibration, different known weights were placed perpendicularly on the sensing area of FSR one by one, and the output voltages were recorded (Prakash and Sharma 2020). The applied loads were converted into force in newton, and the final force vs. voltage curve was drawn.

Plant model description

The prosthetic hand model mainly comprises of actuators coupled to the fingers. The finger-object interaction and finger extension due to elastic force are considered as a disturbance acting over the actuator. Therefore, for such a system, there is a need to design a robust controller that has a disturbance rejection capability. The appropriately tuned controller

Fig. 3 **a** PDMS encased structure of FSR, **b** voltage divider circuit for producing an output of FSR, **c** positioning of FSR sensor at fingertip



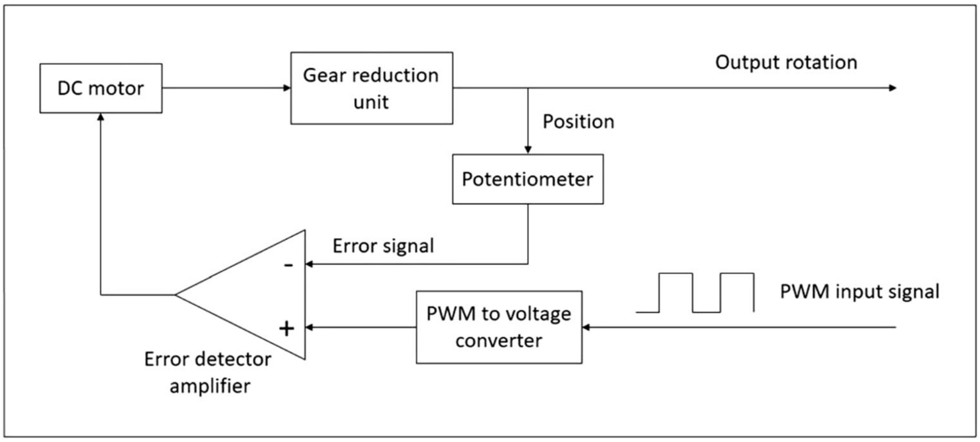


Fig. 5 Block diagram for RC servomotor

provides immediate disturbance rejection capability to the system (Bordignon and Campestrini 2018).

The actuator used is a radio control (RC) servomotor consisting of a DC motor, gear reduction unit and a position feedback controller. Figure 5 shows the block diagram of the RC servomotor describing its operation. The DC motor can be represented by a second-order continuous linear transfer function, in which each pole describes the features of the system. Also, the position controller of the servomotor is hypothetically regarded as proportional one, and the number of teeth of gears can determine the gear ratio (Wada et al. 2009).

Plant model transfer function using system identification

An experiment was performed for determining the continuous transfer function of the servomotor in real-time using the LabVIEW software platform and NI ELVIS II+ (a hardware interface from national instruments). RC servomotors usually have only three input pins, i.e. ground, V_{cc} and PWM (for receiving input signal). Therefore, a separate analog output pin was taken out from the feedback potentiometer of the servomotor for determining its current position. The PWM pin of the servo was attached to the analog output port of the ELVIS board, whereas the newly made analog output pin was connected to the analog input port of the ELVIS.

A LabVIEW program was developed in which PWM output at a variable duty cycle was provided to the servomotor for the real-time estimation of its transfer function using the analog output. The obtained transfer function of the plant is described in Eq. (1).

$$T(S) = \frac{9.29676}{9.31129S^2 + 2.41187S + 1} \tag{1}$$

Control system design

A closed-loop control architecture shown in Fig. 6 was proposed for the developed myoelectric hand prosthesis. Servomotor, coupled to prosthetic hand fingers, is considered as the plant whose transfer function has already obtained by the experimental procedure. The activities like the extension of fingers and finger-object interaction can be assumed as the disturbance to the system. The reference input is the EMG linear envelope of 0–5 V, whereas the prehension force measured by the tactile sensor in 0–5 V is the feedback signal, and the error signal to the controller is the difference between these two signals. The feedback signal basically gives the indication about the angular position of the

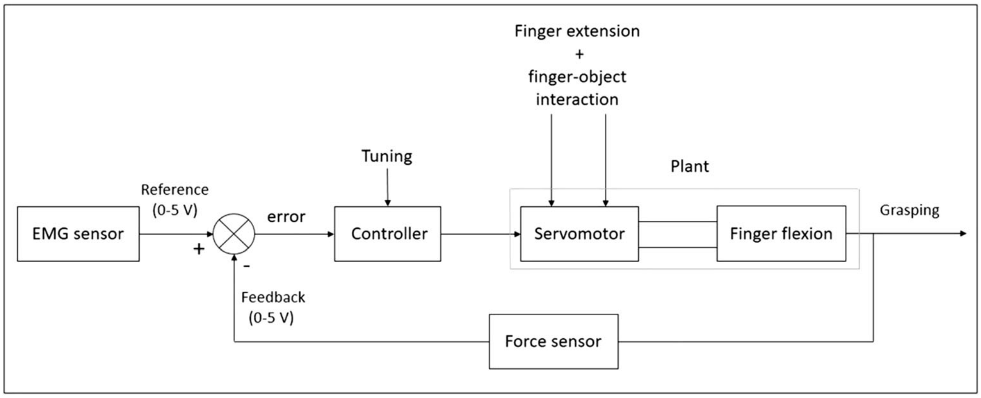


Fig. 6 Proposed control system architecture for the developed myoelectric hand prototype

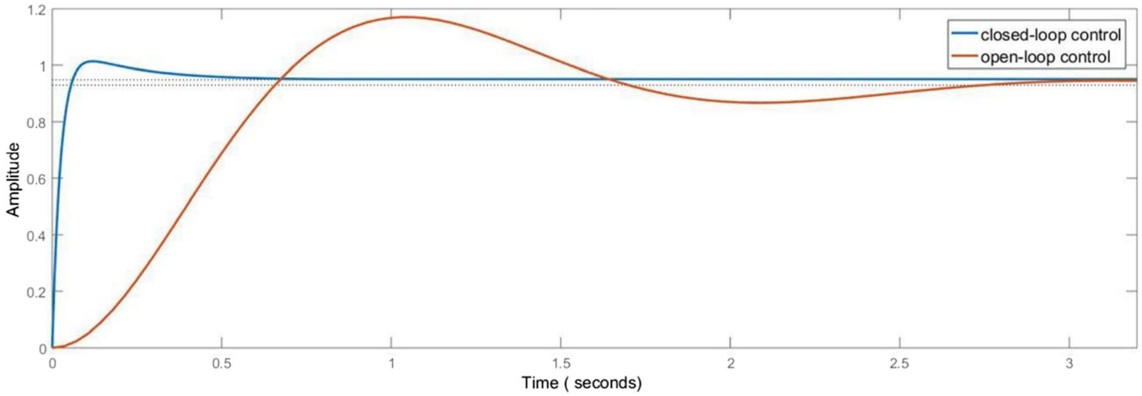


Fig. 7 Step response obtained for open and closed-loop control system

finger interacted with the object. Since more is the flexion of fingers, more will be the generation of prehension force to grasp objects. The designed controller is appropriately tuned to produce the desired command for the flexion of prosthetic hand fingers. The output of the controller is the PWM signal, which provides angular rotation to the servomotors.

The step response characteristics calculated for the open-loop transfer function of the plant (i.e., Eq. 1) showed a higher overshoot of 25% and settling time of 2.5 s. In the myoelectric control system, the values of these two parameters should be optimum (i.e., as less as possible) to achieve fast and precise grasping of objects. The presence of higher overshoot in such a system provides a large grasping force that may lead to the crushing of the delicate objects (Sono and Menegaldo 2009). Also, the large settling time affects the speed of operation of the prosthetic hand. As the open-loop based control for hand prosthesis is unable to provide accurate grasping of objects due to the presence of high overshoot in the system. A tactile feedback system with a proportional-derivative (PD) controller was introduced to attain optimum time response characteristics for the precise grasping of objects. PD force position controller with voluntary EMG signal and force feedback loop minimizes the error between the actual muscular contractile force and the measured prehension force of the fingers (Bahrami Moqadam et al. 2018).

Results

Output response

Using the open-loop transfer function of the plant, a closed-loop proportional-derivative (PD) controller was designed and tuned manually on MATLAB to obtain the desired response for the developed prosthetic hand system. Figure 7 shows the step response curve for the open-loop and closed-loop control system, while Table 1 describes the performance parameters for both the system.

Real-time implementation of the designed control system

The designed closed-loop control system was further implemented for real-time controlled operation of a myoelectric prosthetic hand. The on-board microcontroller in hand drives the servomotor with PD position control. It interfaces with the EMG sensor, the tactile sensor and the PC with the written algorithm. The microcontroller receives the EMG signal as a reference input, grasping force as feedback, and produces the PWM signal as a controlled output to the actuator.

Prosthetic hand trial on subjects

The myoelectric hand trial was done on five transradial amputees for the basic grasping of the objects. The subjects were selected based on their level of amputation, availability and the ease of attachment of the EMG sensor on their residual limb. The details of amputees with their type and reason for amputation are mentioned in Table 2. The sensor was attached to the remaining forearm stump of amputees for capturing the EMG signal. Figure 8 shows the grasping of four different shaped objects, i.e. (a) sponge ball, (b) plastic container, (c) cylindrical glass and (d) conical glass with the prosthetic hand using EMG signal from an amputee.

Table 1 Step response characteristics for different control systems

Performance parameters	Rise time (s)	Overshoot (%)	Settling time (s)
Open-loop control	0.445	25	2.57
Closed-loop control with PD controller	0.253	4.55	0.382

Table 2 Details of amputees participated in prosthetic hand trials

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

During grasping of four different shaped objects, the reference EMG signal, the feedback signal and the error signal were captured in real-time using data acquisition (DAQ) device. This was done to know the nature of reference input, feedback and error signal (i.e. input to the controller). Figure 9 shows the real-time response of reference EMG signal, feedback signal (prehension force), and the error signal during grasping of four different shaped objects for an amputee.

Furthermore, the prehension force of hand fingers was measured during grasping four different objects using output from the tactile sensor. The measured prehension force, while grasping of objects for each subject are listed in Table 3.

Discussion

It is quite clear from Fig. 7 and Table 1 that the proposed closed-loop control system showed improved parameters (i.e. rise time, settling time and overshoot) than the open-loop system. The lower settling time will offer the faster operation of prosthetic hand fingers to close the hand. On the other hand, the lesser value of overshoot will promote the precise grasping of objects with the hand fingers.

During the trial, subjects using their distinct strengths of EMG signal were able to grasp different objects precisely with the hand. Each grasping activity was repeated five times for every subject. All the subjects revealed that the developed myoelectric hand was able to provide precise grasping of objects using the user's intention, i.e., quite similar to that of a natural hand. Furthermore, there was no report of fatigue from any of the user as each trial was conducted only for 1 h.

The response time, i.e. full closing/opening time of the myoelectric hand, was experimentally determined from the recorded video of hand operation. The obtained closing/opening time of hand was 400/450 ms comparable to that of Ottobock sensor hand (i.e. 300/300 ms), which is considered as the fastest commercial hand. Moreover, the survey reports revealed that 300–400 ms could be the acceptable closing time for the myoelectric prosthesis (Belter et al. 2013; Englehart and Hudgins 2003; Farrell and Weir 2007).

In Fig. 9, reference input is EMG signal voltage, which is proportional to the force of muscle contraction, whereas the feedback signal is the measured prehension force (in terms of voltage) during grasping. So depending on the measured prehension force, the user can apply the muscular contractile force (i.e. the intensity of EMG signal) for precise grasping of objects. Additionally, if we see the output response, it is the angular position of the finger, which is being measured in terms of force with the help of a tactile sensor during object-finger interaction. Therefore one response is position while the other is force position whose comparison is possible. The error signal produced is input to the PD controller for producing the desired output response, which is further converted to a PWM signal for driving servomotor.

From Table 3, it can be concluded that the measured prehension force varies with the shape and weight of objects. For

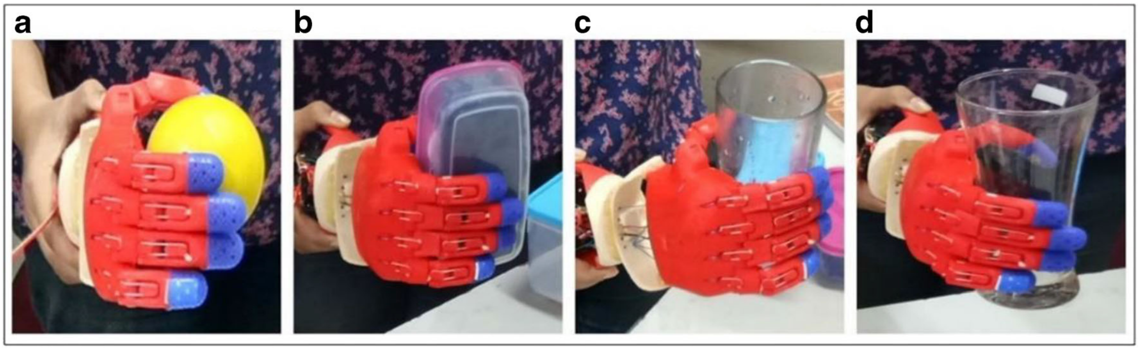


Fig. 8 Grasping of four different shaped objects: **a** sponge ball, **b** plastic container, **c** cylindrical glass, **d** conical glass

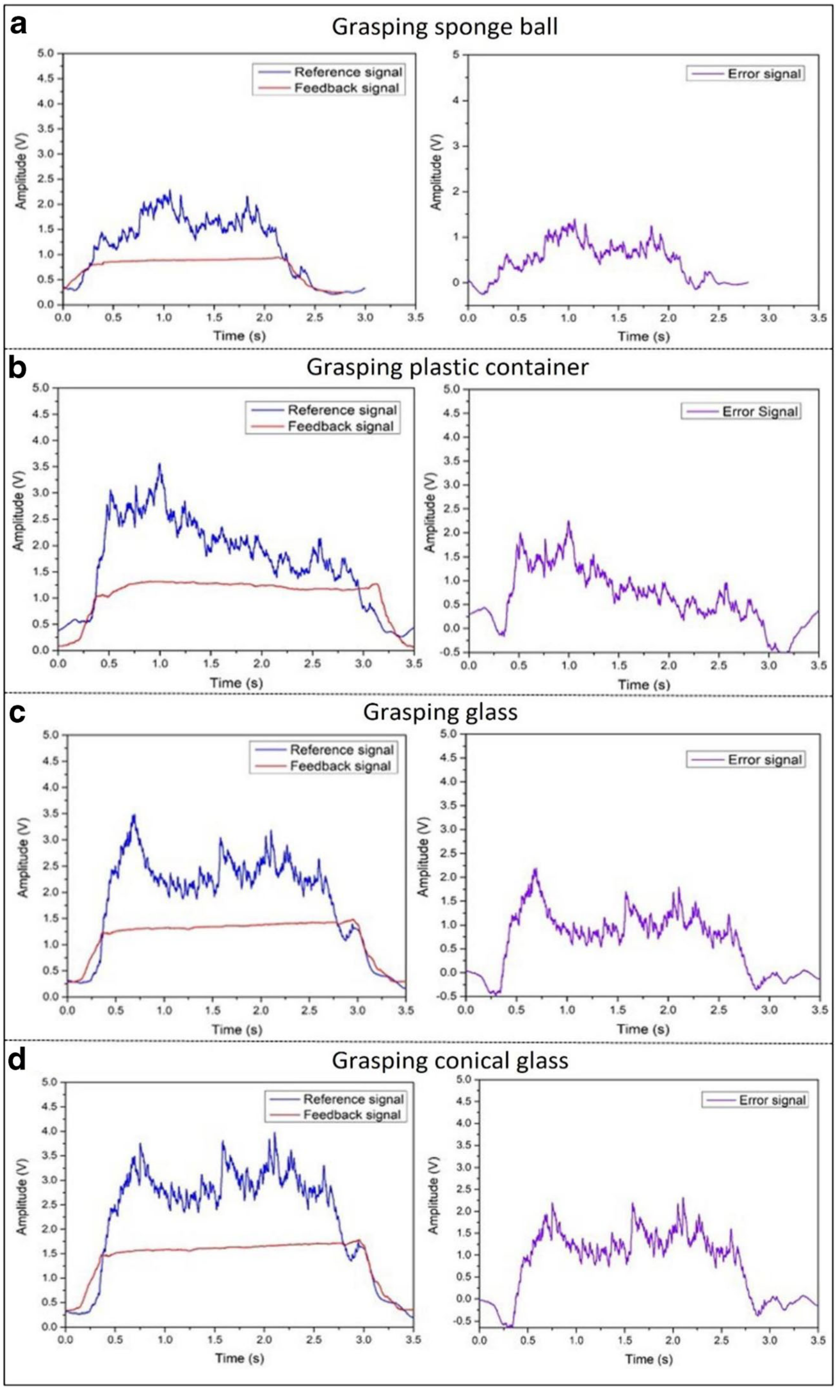


Fig. 9 Real-time response of EMG signal, feedback signal and error signal during grasping of four different shaped objects

Table 3 Prehension force for different subjects

Subject	Prehension force (N)			
	Grasping sponge ball	Grasping plastic container	Grasping glass	Grasping conical glass
1	0.85	1.42	2.12	2.54
2	0.91	1.45	2.21	2.65
3	0.84	1.44	2.15	2.62
4	0.92	1.46	2.20	2.70
5	0.90	1.43	2.16	2.64

Table 4 Comparison of developed myoelectric hand system with commercial hand

Parameters	Developed hand	Ottobock sensor hand
Mass	350 g	460 g
Material	Polylactic acid (PLA)	Silicone
Fabrication technique	3D printing (FDM)	Moulding
Size	175 × 85 × 35 mm	184 × 80 × 40 mm
DOF	3	1
Number of actuators	2	1
Actuation method	Dc motor-tendons	Dc motor-worm gear
Control scheme	Proportional-derivative	Proportional
Feedback	Force	Force
Full closing time (response time)	400 ms	300 ms
Battery	3.7 V, lithium-polymer, 2000 mAh	7.2 V, lithium-ion, 2200 mAh
EMG sensor	Single-channel	Single-channel
Price in the commercial market	Prototyping cost ($50)	$42,000

grasping stiffer and heavier objects, larger prehension force is required, whereas softer and lighter objects require smaller prehension force. However, the prehension force values for grasping distinct objects were observed approximately the same for all the subjects.

Conclusion

This paper presents a simple and low-cost system for controlling the prehension force of a custom-made myoelectric hand prosthesis. A 3D-printed hand prototype was prepared and intrinsically actuated using two servomotors. A sensitive tactile sensor was installed at the fingertip for the online measurement of prehension force. A closed-loop proportional-derivative (PD) based control system was designed with the force feedback mechanism and EMG signal as a reference, using the determined transfer function of the plant, i.e. servomotor. The developed system showed improved time-response parameters, i.e. overshoot, rise time and settling time as compared to the open-loop system. Higher overshoot in the hand prosthesis system leads to the production of the larger prehension force of the fingers, whereas the rise and settling time determines the speed of operation of the prosthetic device. The designed closed-loop control system with the PD scheme was able to produce a desired response for the myoelectric prosthesis system.

Further, the designed control system was realized in the developed prosthetic hand prototype. The hand was tested on five different transradial amputees for the dexterous grasping operation of different shaped objects. The amputees using their EMG signal were able to grasp different objects with control on prehension force delicately. The main purpose of designing the control system was to achieve precise grasping of objects using the prosthetic hand (i.e. avoiding its slippage as well as damage of object). The hand prosthesis was able to produce smooth, faster and intuitive grasping operation, which shows its capability in performing activities of daily livings.

Real-time measurement of prehension force was made concerning the EMG signal during the grasping of different objects. The measured values indicated the required prehension force for grasping different objects.

Table 4 describes the comparison of developed myoelectric hand system with a commercial hand in terms of various parameters.

The hand prototype, with the designed control system, offers several advantages like simple as well as low-cost setup, provides faster and delicate grasping operations, delivers natural hand like control and does not require prior training. However, the main drawback of the developed myoelectric hand system is its limited number of grip patterns. The disadvantage can be overcome by introducing pattern recognition-based control in conjunction with the designed closed-loop control system.

The inclusion of such an efficient and low-cost system will enhance the grasping capability of myoelectric hand prosthesis system, leading to an increase in the acceptance rate among amputees.

Acknowledgements The authors would like to thank the Design Innovation Centre, Indian Institute of Technology (BHU) for funding this project.

Funding information This research work was funded by Design Innovation Centre, Indian Institute of Technology (BHU).

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 included in the study.

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