Chapter 1

Introduction

1.1 Amputation

Amputation is the physical loss of a limb or its portion by trauma, cancerous tumor in muscles and bones, vascular disease, infection, etc. It severely worsens the quality of life by restricting body functionality. Loss of limbs has an immense effect on the amputee's body image and work productivity (Sahu et al. 2016). Patients suffering from limb loss are classified as upper and lower limb amputees. Upper-limb amputation refers to the missing arm or a part of the arm, while lower-limb amputation relates to the lost leg or its part.

1.1.1 Upper-limb amputation

Upper-limb amputation is further categorized into two main types: transradial and transhumeral. The missing limb above the elbow is known as transhumeral amputation, which includes shoulder disarticulation, forequarter, and elbow disarticulation. On the other hand, transradial amputation refers to the loss of limb below the elbow, which includes hand/wrist disarticulation and partial hand, i.e., transcarpal amputation (Maduri and Akhondi 2019; Ovadia and Askari 2015). This thesis work mainly targets below-elbow amputations. Figure 1.1 describes the various levels of upper-limb amputation, in which the lost limb below the elbow is highlighted as the focus of this research.

1.1.2 Amputation scenario

There are approximately 15 million amputees across the world, out of which 30% are upper-limb amputees. The majority of these amputees belong to developing countries. Moreover, nearly 60% of the upper-limb amputees from low-income countries are transradial ("World Report on Disability," 2011; Slade et al. 2015). In India alone, around 0.75 million upper limb amputees are there, out of which 0.3 million are transradial. The main reason for below-elbow amputation in developing countries is trauma, vascular disease, and malignancy (Hamner, Narayan, and Donaldson 2013). Figure 1.2 gives the statistics of upper-limb amputees residing in the world, developing countries such as India.

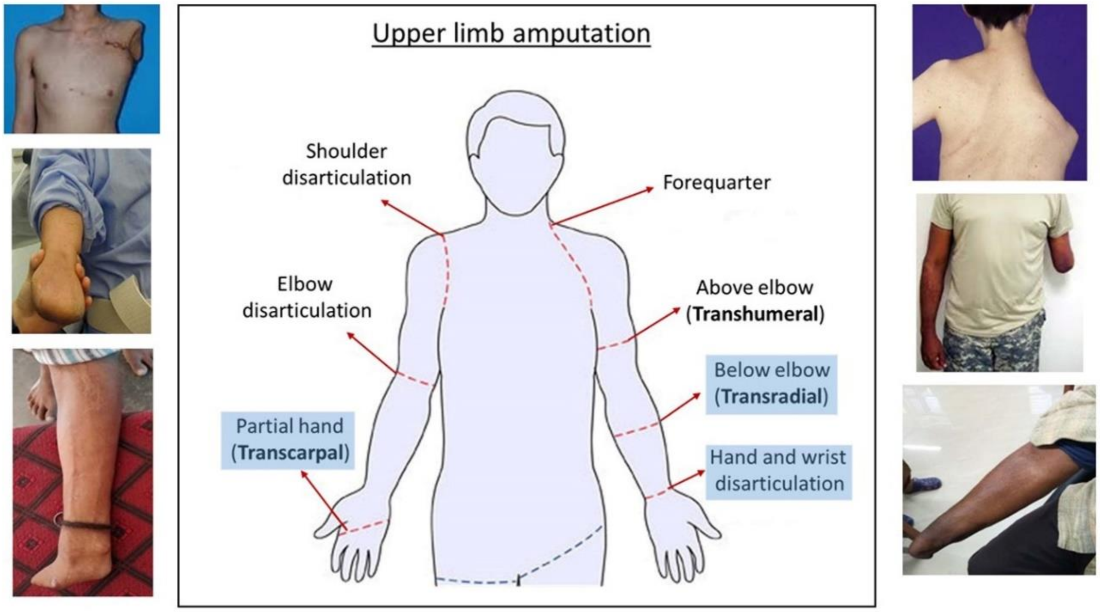

Figure 1.1 Different types of upper-limb amputation.

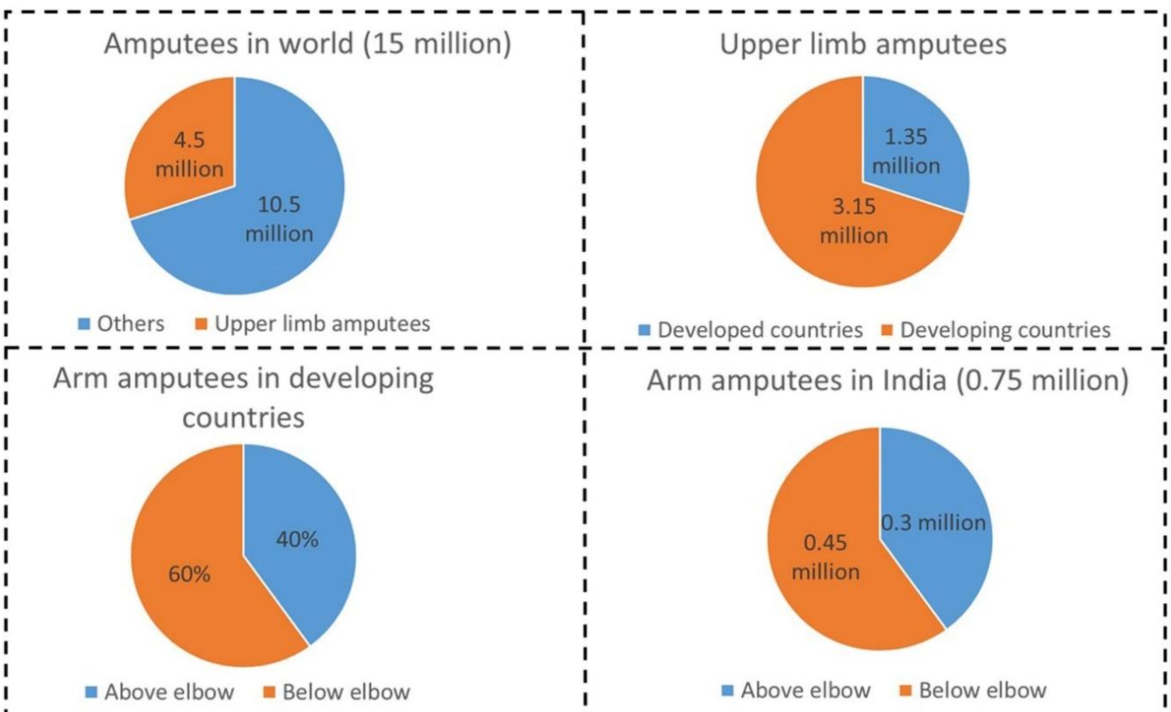

Figure 1.2 Statistics of amputees around the world.

1.2 Prosthesis

A prosthesis is an artificial device that substitutes or aids a portion of the body or an organ, which may be lost by trauma, infection, or a disorder existing at birth. The prosthesis is considered the primary solution to amputations, as these are capable of reinstating close to normal functions of the missing body part. Limb prosthesis or artificial limb is classified as an upper and lower-limb prosthesis, providing alternatives to different types of arm and leg amputations (McGimpsey et al. 2008).

1.2.1 Upper-limb prosthesis

Based on the level of amputation, the upper-limb prosthesis is classified as transradial, transhumeral, and transcarpal. A prosthesis meant for above-elbow amputation is transhumeral, whereas transradial prosthesis refers to the artificial limb for below-elbow amputation. And transcarpal prosthesis is an artificial substitute used for partial hand amputees (McGimpsey et al. 2008; Ovadia and Askari 2015). Figure 1.3 illustrates the types of upper-limb prosthesis based on amputation level.

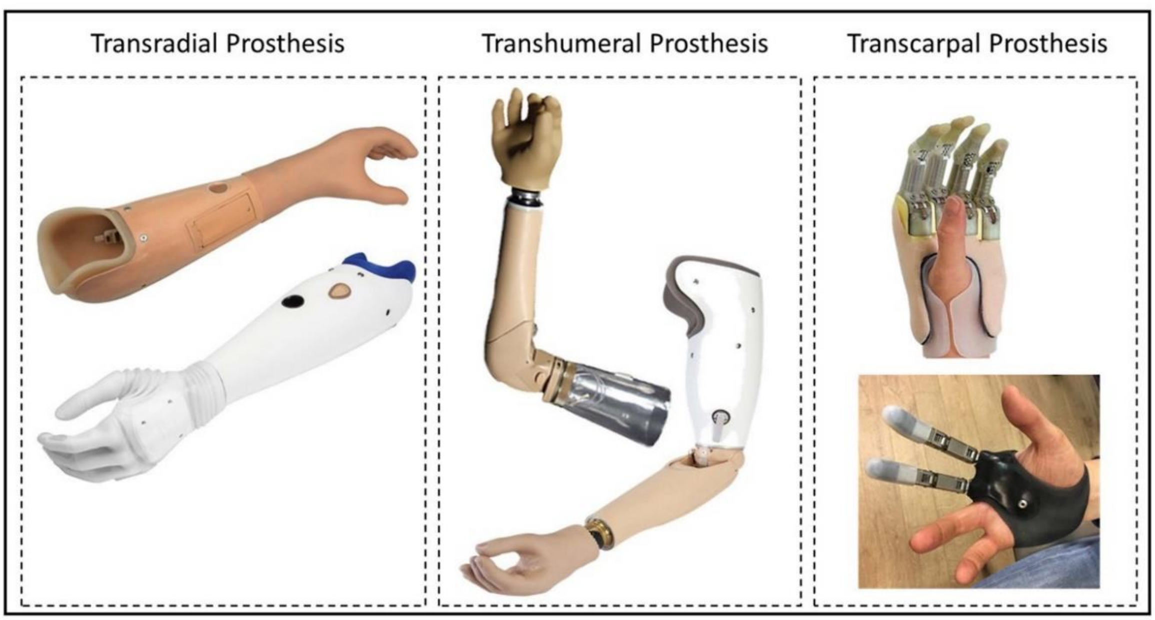

Figure 1.3 Different types of upper-limb prosthesis.

On the basis of functionality, the upper-limb prosthesis is again classified as body-powered and externally powered (Geethanjali 2016). The classification is depicted in Figure 1.4.

1.2.1.1 Body-powered prosthesis

The body-powered prosthesis consists of hooks that 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 (Figure 1.4). The main disadvantage of the body-powered prosthesis is it requires considerable body power to move the prosthesis (puts more strain on the amputee's body), unnatural grasping of an object and creates discomfort while wearing (Uellendahl 2017).

1.2.1.2 Externally powered prosthesis

Externally powered prostheses, on the other hand, are battery operated and are controlled by bioelectric signals like electromyography (EMG), electroencephalography (EEG), targeted muscle reinnervation (TMR), electrocorticography (ECOG) and electrical signals from other sensors and switches (Cordella et al. 2016). EMG and EEG techniques are mostly preferred for controlling prostheses because of their non-invasiveness, whereas ECOG and TMR are invasive techniques (Cheesborough et al. 2015; Fifer et al. 2012). EEG can be a decent option for directly controlling prosthetics with brain signals. However, these signals have several limitations, i.e., low reliability, low accuracy, low user adaptability, low data transfer rate, and complex acquisition setup (AL-Quraishi et al. 2018).

EMG can directly reflect the user's motion intention through muscular contractions, which can be effectively utilized to control prosthetics intuitively. Moreover, EMG requires an easy acquisition setup and is reliable compared to other means (Jiang et al. 2014; George et al. 2020; Milosevic, Benatti, and Farella 2017).

The electrical signals directly from the sensors and switches may provide unnatural control to prosthetics. However, there are some biomechanical techniques such as force myography (FMG), optomyography (OMG) and mechanomyography (MMG), which can measure mechanical muscular contractions by the indirect use of force sensors, optocouplers and accelerometers (Cordella et al. 2016). The detection of mechanical activity of muscle can also be applied to control the prosthetic devices in the same manner as utilizing EMG (i.e., detecting electrical activity of muscle) (Esposito et al. 2018).

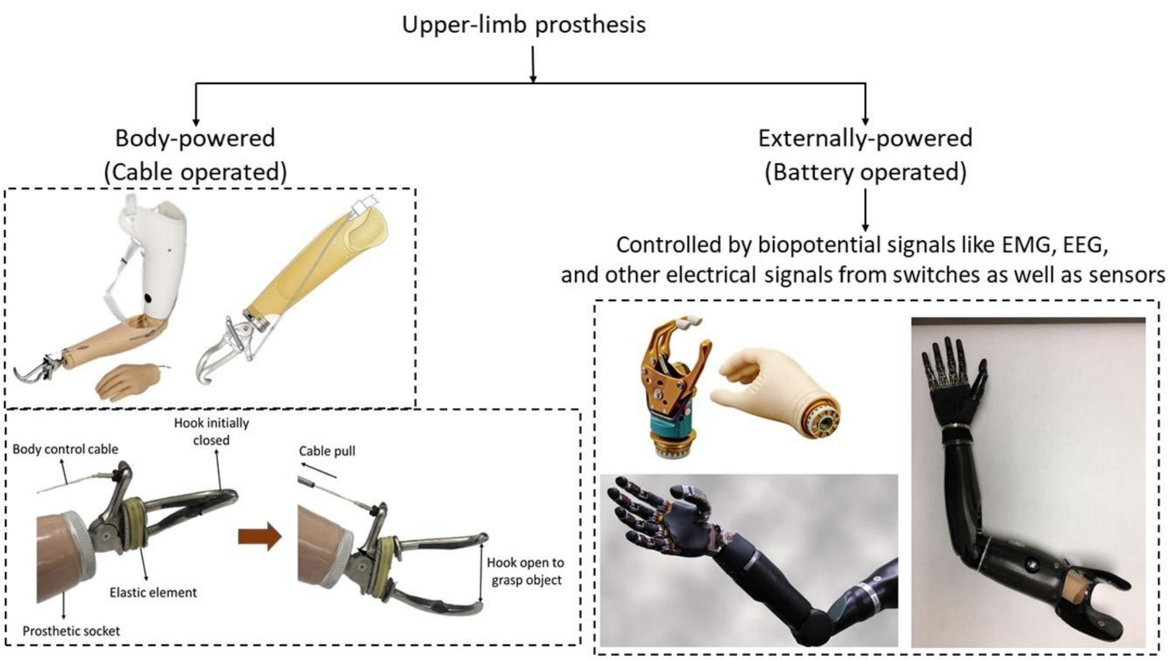

Figure 1.4 Different types of upper-limb functional prosthesis.

1.3 Electromyography

Electromyography (EMG) is an approach for measuring the electrical activity of muscles. Contraction or activation of a particular muscle produces electrical changes inside the muscle, which can be detected by placing electrodes on the skin. The electrical signal produced is referred to as EMG or myoelectric signal. The EMG signal is mainly the voltage variation resulting from the flow of ionic current across the muscle membranes when muscle cells are activated electrically or neurologically. Therefore, by analyzing the EMG signal, the underlying biological process of muscles can be studied. (Day, 2002; Hefftner, Zucchini, and Jaros 1988).

The central nervous system in the cerebral cortex part of the brain controls the motor nerve cells to produce neural impulses. For performing specific muscle activity, neural impulses from the motor cortex are transmitted to motor neurons and then to muscle fibers (motor unit). Each fiber generates its own motor unit action potential and whose continuous activation produces motor unit action potential trains (MUAPTs). Individual MUAPTs of each fiber are superimposed to give the EMG signal (Ali, Albarahany, and Liu quan 2012; Jamal 2012). Figure 1.5 describes the principle of generation of EMG signal.

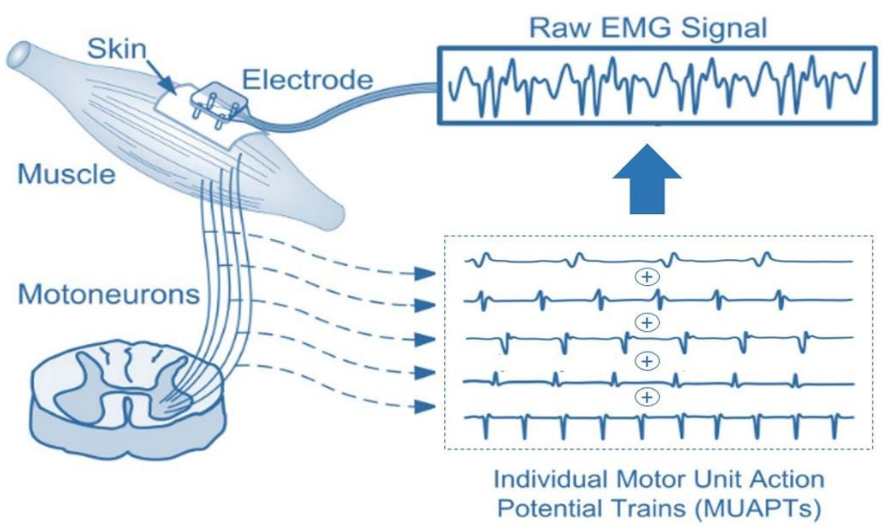

Figure 1.5 Generation of the EMG signal.

EMG signals are measured either by placing electrodes on the muscle surface or by inserting needles invasively within the muscle. Surface Electromyography (sEMG) is usually preferred because it is a non-invasive technique in which electrodes are placed directly on the skin in conjunction with the muscle surface of the human subject (De Luca 1997). Figure 1.6 shows the placement of electrodes on the skin surface for the measurement of the EMG signal. It is a three-electrode configuration in which two electrodes are placed on the skin (above the target muscles), while the third electrode is used as a reference that can be positioned at any electrically un-coordinated point. The signal picked by the two target electrodes is given to the instrumentation amplifier to produce a differential EMG signal (Jamal 2012).

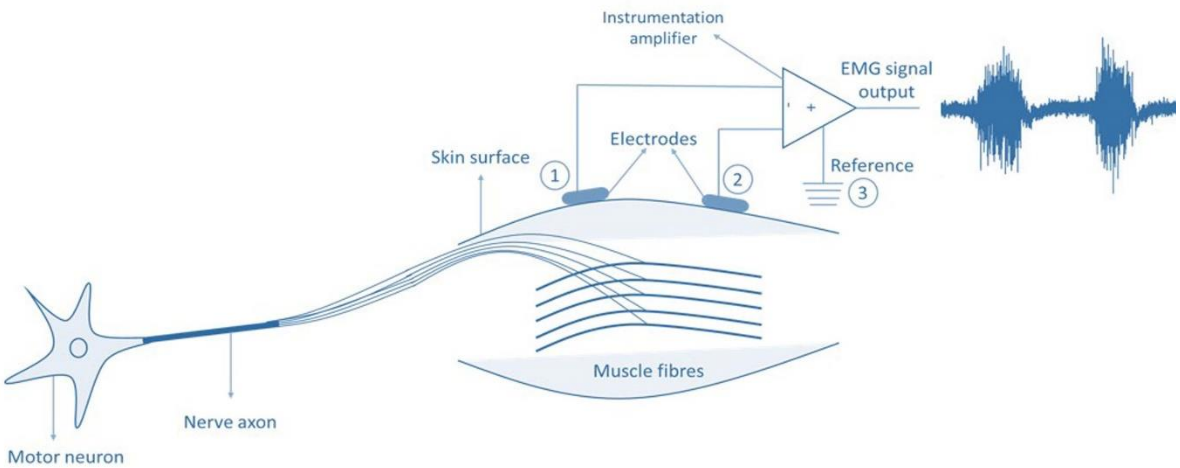

Figure 1.6 Placement of electrodes for performing surface electromyography.

The amplitude of the raw EMG signal lies in the range of 0-10 $mV_{p-p,}$ whereas its frequency exists between 0-1000 Hz, in which most relevant information is contained within 10-350 Hz. The measurement system typically consists of electrodes and proper signal conditioning circuitry for correctly detecting the EMG signal patterns (Shobaki et al. 2013; Day, 2002). The electrodes sense the raw EMG signal while the signal conditioning circuitry consisting of amplifiers and filters converts the raw signal into a more usable form (i.e., a signal in the suitable frequency range and with sufficient amplitude) (Jamal 2012; Pylatiuk et al. 2009). Figure 1.7(a) and 1.7(b) describes two different EMG measurement systems.

The EMG signals have significant application in the diagnosis of neuromuscular disorders and controlling assistive devices. Nowadays, EMG has become a prominent source of control for the upper-limb prosthesis (J. Liu and Zhou 2013; Tavakoli, Benussi, and Lourenco 2017; Pancholi and Joshi 2018).

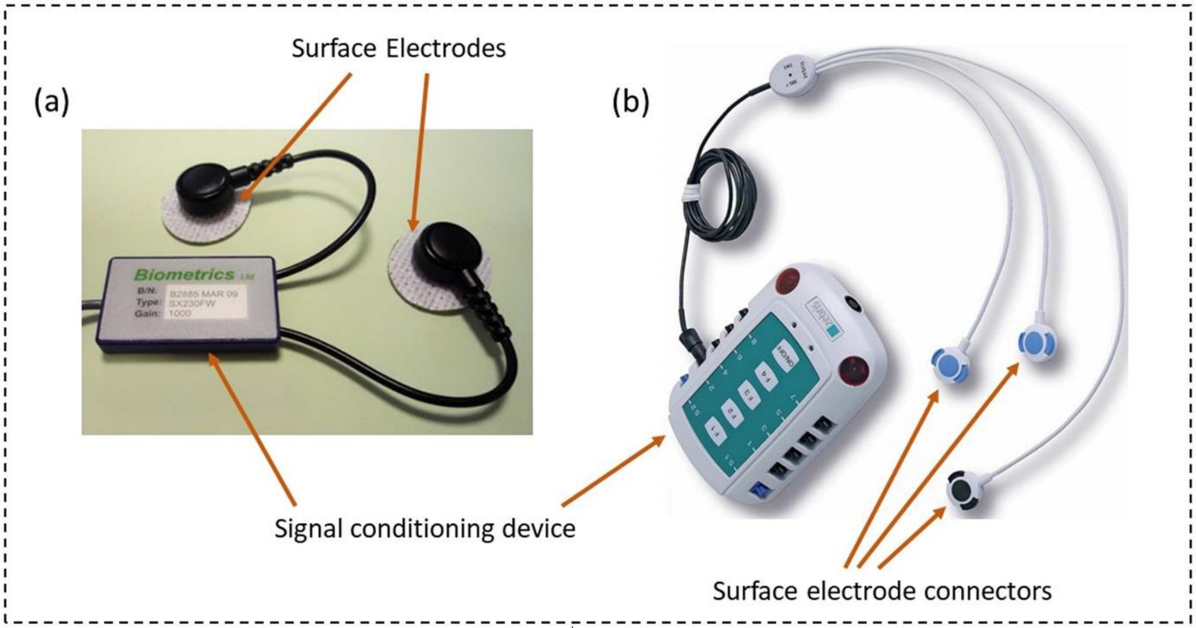

Figure 1.7 Description of EMG measuring devices.

1.4 Myoelectric prosthesis

A myoelectric prosthesis is a type of externally powered prosthesis that utilizes EMG signals from the residual upper-limb of patients to operate the terminal device, using a suitable control strategy. Figure 1.8 depicts the chief components present in a myoelectric hand prosthesis. It essentially includes (1) one or more EMG sensors, (2) hand assembly or terminal device consisting of actuator unit, (3) microcontroller unit, (4) battery, and (5) socket (Parker, Englehart, and Hudgins 2006; Asghari Oskoei and Hu 2007). EMG sensors extract the bioelectrical signals generated by the muscle contractions of the residual upper-limb. The hand assembly has a mechanical linkage with the actuator unit through tendons, which allows it to perform functions. The microcontroller unit receives the EMG signals from the sensors and converts them to control commands for driving the actuator unit. The battery delivers the primary power source to drive the whole prosthesis system. The socket connects the hand prosthesis with the residual upper-limb of the user. It also provides significant strength to a prosthesis for holding various objects (Pasquina et al. 2015).

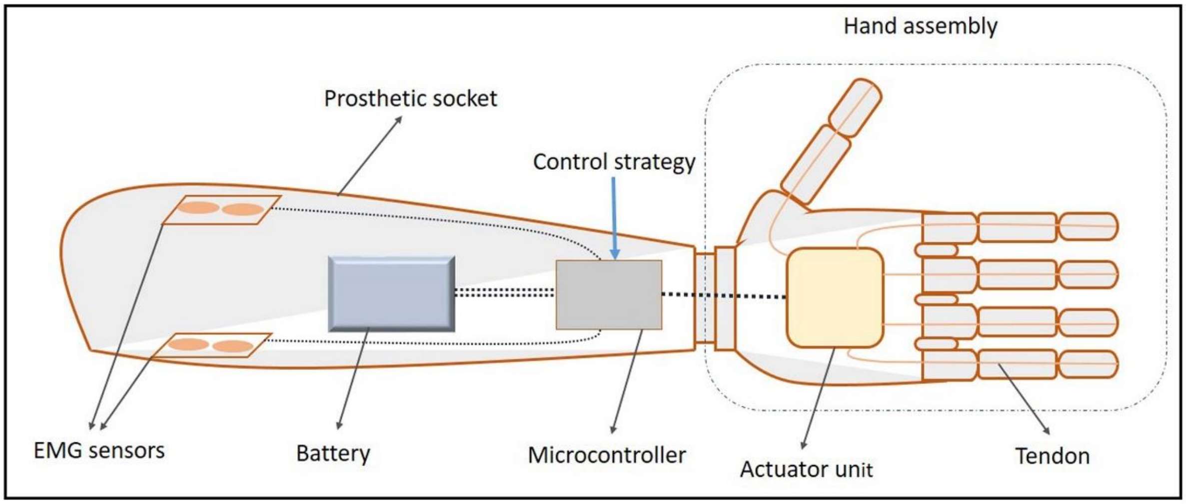

Figure 1.8 The various components of a myoelectric prosthesis.

1.5 Myoelectric control system

Myoelectric control system (MCS) principally converts the EMG signal features into control commands for driving the prosthetic device. The MCS utilizes the valuable information from the amputee's residual limb regarding muscular contractions to provide control signals to a prosthesis for performing operations (Asghari Oskoei and Hu 2007; Geethanjali 2016). The role of the MCS in a myoelectric prosthesis is very significant since it decides how the actuation of the terminal device will be executed as per the input EMG signal. MCS is categorized as pattern recognition (PR) and non-pattern recognition (NPR) based control system. In the NPR based control scheme, the input EMG signals are directly mapped into control instructions using a suitable algorithm. Such a control scheme can only provide limited degrees-of-freedom (DOF) movement to the prosthesis. Proportional and threshold controls are the two conventional and commonly used NPR based approaches (Lenzi et al. 2012; Fougner et al. 2012).

On the other hand, PR based control utilizes complex classification techniques for translating the EMG signal patterns into some predefined hand activities, which are further transformed into control commands using NPR based strategy. PR based schemes can provide a multi-DOF operation to prosthetic hands with fine grasping (Englehart et al. 1999; Ahsan, Ibrahimy, and Khalifa 2011). Figure 1.9 describes the two different types of MCS.

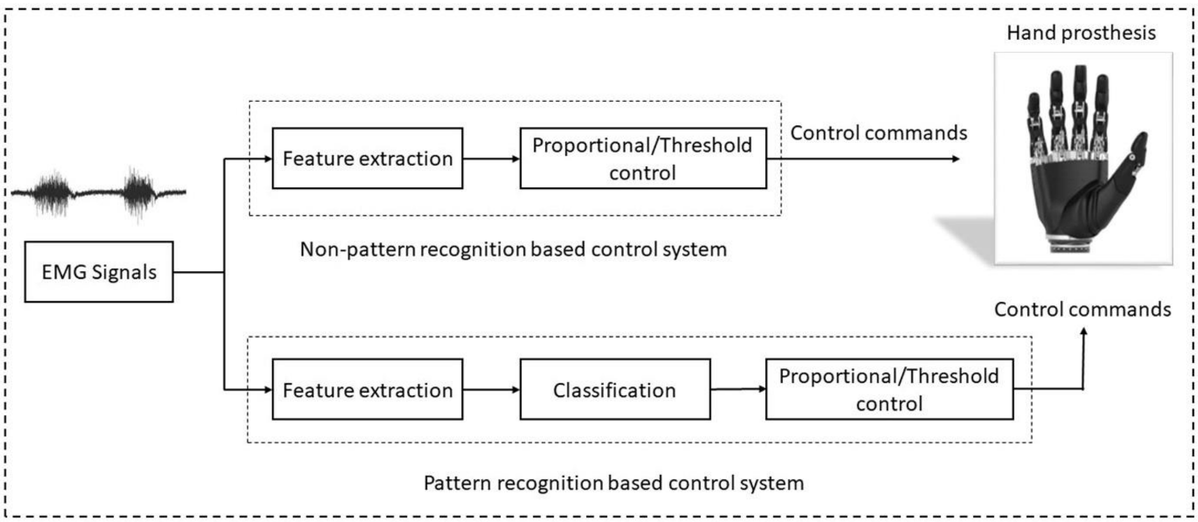

Figure 1.9 Types of a myoelectric control system.

1.6 Literature survey

Several efforts have been made regarding the design and development of externally powered prostheses, which can mimic a real arm in terms of intuitiveness, functionality, and appearance. Some of the work carried out by various researchers has been described below. Table 1.1 presents a comparative chart depicting important features of the hand prostheses developed by some of these researchers.

Manus hand was developed having four basic grasping modes with only two actuators. The intrinsic actuation scheme was implemented for the hand, i.e., the location of actuators inside the hand palm (Figure 1.10a). The hand receives the control signal from offline classified EMG signals. The hand was able to produce a maximum grasping force of 60 N at the fingertip. The size of the prototype was 20% larger than the human hand, whereas the weight of the hand was 1200 g (Pons et al. 2004).

A hand prototype was designed with three fingers intrinsically actuated by four DC motors through a special under-actuated transmission. The hand was capable of producing a maximum grasping force of 35 N. The overall weight of the hand was 320 g. The hand was mainly designed for robotic applications (Zollo et al. 2007).

Dalley et al., 2009 designed a multifunctional hand providing eight different grip patterns utilizing multi-channel offline EMG signals. The hand prototype was developed, consisting of 16 joints driven by five independent actuators (with an intrinsic actuation scheme). The hand operates in an open-loop force control configuration and provides a maximum force of 50 N through the fingertips. The weight of the hand was 580 g, whereas its closing time was 0.48 s (Dalley et al. 2009).

Smarthand was developed with five fingers actuated by four motors, providing 16 degrees of freedom (Figure 1.10b). EMG pattern recognition based control scheme was implemented to obtain individual finger movement and five different grasping patterns with the hand. It integrates 32 force, position, and tactile sensors for automatic grasp control and sensory feedback to the subject. The weight of the hand was 530 g, and its full closing time was 1.5 s. The hand was capable of holding objects up to 3.6 kg (Cipriani, Controzzi, and Carrozza 2011).

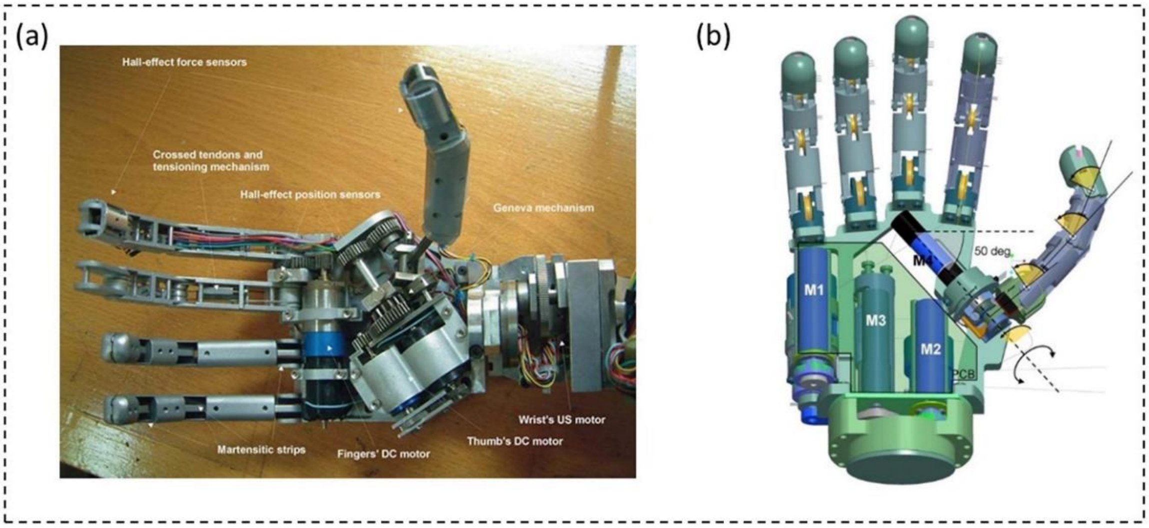

Figure 1.10 (a) Manus hand intrinsically actuated using two DC motors(Pons et al. 2004), (b) Smart hand actuated by four motors (Cipriani, Controzzi, and Carrozza 2011).

Modular prosthetic limb was designed, which receives control inputs from neural interfaces (i.e., targeted muscle reinnervation) and uses various tactile sensors for sensory feedback (Figure 1.11a). The hand was intrinsically actuated using fifteen motors, which could provide up to 22 degrees of freedom (DOF) depending on the level of amputation (Johannes et al. 2011).

Southampton hand fingers were actuated using six electrical motors, of which two for the flexion/extension and rotation of the thumb, whereas the remaining four motors for individual finger flexion/extension (Figure 1.11b). The weight of the hand was 400 g, and it was capable of producing a grasping force of 100 N at the fingertips. EMG signals controlled the hand operation from the implantable six-channel myoelectric sensors and the static and dynamic force sensors installed at fingertips (Vasluian et al. 2014) (Kyberd et al. 2001).

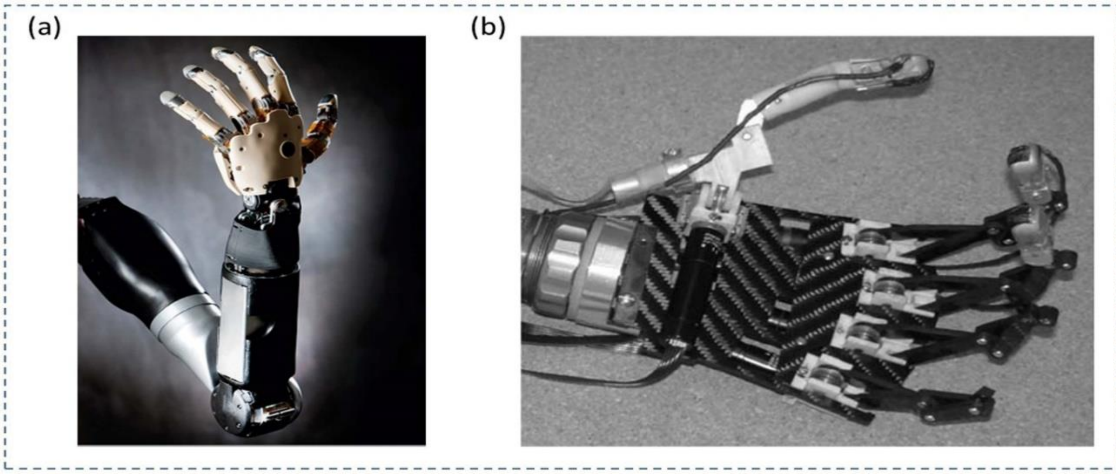

Figure 1.11 (a) Modular prosthetic limb intrinsically actuated by fifteen motors (Johannes et al. 2011), (b) Southampton hand-activated using six motors (Vasluian et al. 2014; Kyberd et al. 2001).

Deka arm from the defence advanced research project agency (DARPA) was designed with six grip patterns and four power wrist movements (Figure 1.12a). The arm offered controlled grasping with a combination of inputs from EMG sensors, force-sensitive resistors (FSR), and other inertial measurement units (IMU). The arm consisted of six motors under the intrinsic actuation mechanism, which was able to provide up to 10 DOF with the Radial configuration of the hand. The weight of the hand was 1270 g (Resnik, Klinger, and Etter 2014).

HIT prosthetic hand was developed by Liu et al., which provided four different grasping modes. Each finger of the hand except the thumb was actuated by an individual dc motor, gear head, and tendon (Figure 1.12b). The thumb had a unique design and actuation scheme that delivers curling and extension motion. The hand size was comparable to the normal human hand, and its weight was 350 g. The hand had strong manipulation abilities, which gives individual finger movements and grasping to fulfill the needs of amputees (Y. Liu, Feng, and Gao 2014).

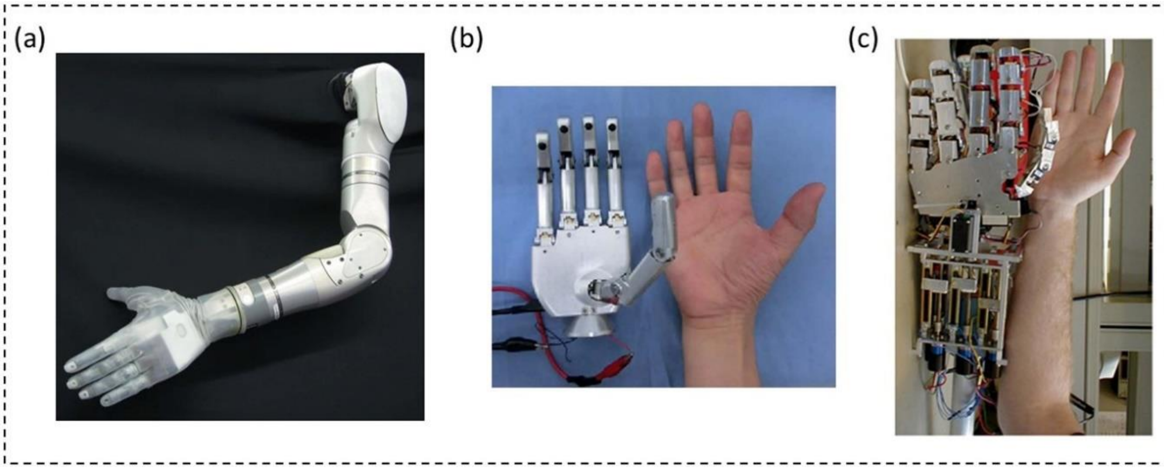

Figure 1.12 (a) Deka arm from defence advanced research project agency (Resnik, Klinger, and Etter 2014), (b) Hit prosthetic hand (Y. Liu, Feng, and Gao 2014), (c) A dexterous hand actuated by seventeen motors (Williams and Walter 2015).

A dexterous hand consisting of 17 actuators was developed utilizing two different actuation schemes for each finger, i.e., single driving tendon and independent joint servomotors (Figure 1.12c). The hand was able to produce various grasping, individual finger motions, and a maximum grip force of 21 N at each fingertip. The hand offered a total of 22 DOF, and its overall weight was 1000 g. The hand was controlled through signals from a glove having potentiometers (Williams and Walter 2015).

An anthropomorphic, single-channel sEMG signal-controlled prosthetic hand was developed consisting of five fingers intrinsically actuated by four independent motors. It utilizes pattern recognition-based myoelectric control for providing eight different gestures to perform activities of daily life. The overall weight of the hand was 500 g, and its full closing time was measured 0.5 s (Wang, Lao, and Zhang 2017).

A 3D printed hand prototype intrinsically actuated by six dc motors was developed, which was capable of providing a total of 6 DOF (Figure 1.13a). It employed multiples sensors (i.e., force sensor, Hall Effect sensor, potentiometer, etc.) to enhance the hand's grasping capability by providing feedback to the controller. The hand utilizes a three-channel EMG system for individual and easy control of each finger (Borisov et al., 2017).

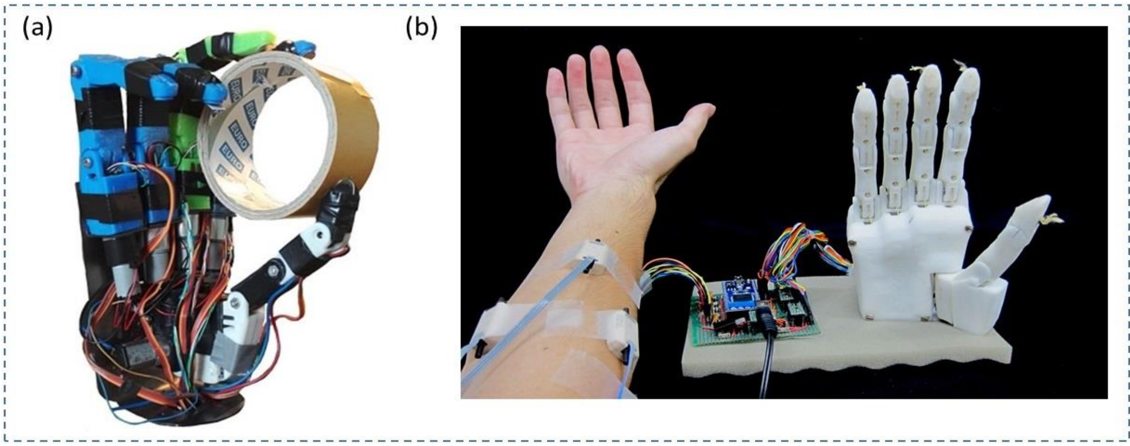

Figure 1.13 (a) An anthropomorphic, single-channel sEMG controlled prosthetic hand (Borisov et al., 2017), (b) 3D-printed myoelectric prosthesis with individual finger operation (Furui et al. 2019).

A 3D printed transradial prosthesis based on a wireless EMG sensor was developed to perform symmetrical grasping of objects. The motor-tendon driven hand fingers could perform three different hand motions (Farooq et al. 2019).

Furui et al. developed a 3D-printed myoelectric prosthesis with individual finger operation (Figure 1.13b). The control was based on the classification of muscle synergy and the biomimetic impedance model, which results in ten different grasping modes of the prosthetic hand. The unique control scheme enabled smooth and intuitive movement of hand fingers similar to that of a human hand (Furui et al. 2019).

Parameters	**Pons et al. 2004**	**Dalley et al. 2009**	**Cipriani et al. 2011**	**Vasluian et al. 2014**	**Resnik et al. 2014**	**Williams et al. 2015**	**Wang et al. 2017**
Weight (g)	1200	580	530	400	1270	1000	500
DOF	4	16	16	16	10	22	16
Actuators used	2	5	4	6	6	17	4
Control signal	Offline classified EMG	Multi-channel offline EMG	Multi-channel EMG	Multi-channel EMG	EMG+FSR+IMU	Glove having potentiometer	Single-channel EMG
Feedback	no	no	sensory	force	no	no	no
Full closing time (s)	-	0.48	1.5	-	-	-	0.5
Max. grip force (N)	60	50	35	100	-	21	-
No. of grip pattern	4	8	5	5	6	-	8

Table 1.1 Prosthetic hands developed by the researchers.

There is a rapid growth in the research and development of biomimetic prosthetic hands in the last few years. The researchers have accomplished a significant amount of work regarding the development of functional hand prostheses, which can provide features like fine grasping, individual finger movement, and prehension force control. However, the end products are still confined to research laboratories only, i.e., their clinical applicability is yet to be realized. The main issues are their functionality, weight, size, unnatural control, operating time, etc., resulting in the rejection of these devices from amputees.

1.7 Market overview

Some commercially available prosthetic hands (Figure 1.14) which can offer features similar to that of the natural hand are described below in Table 1.2 ("Michelangelo Prosthetic Hand"; "Myoelectric Speed Hands"; "I-Limb Quantum"; "Bebionic Hand").

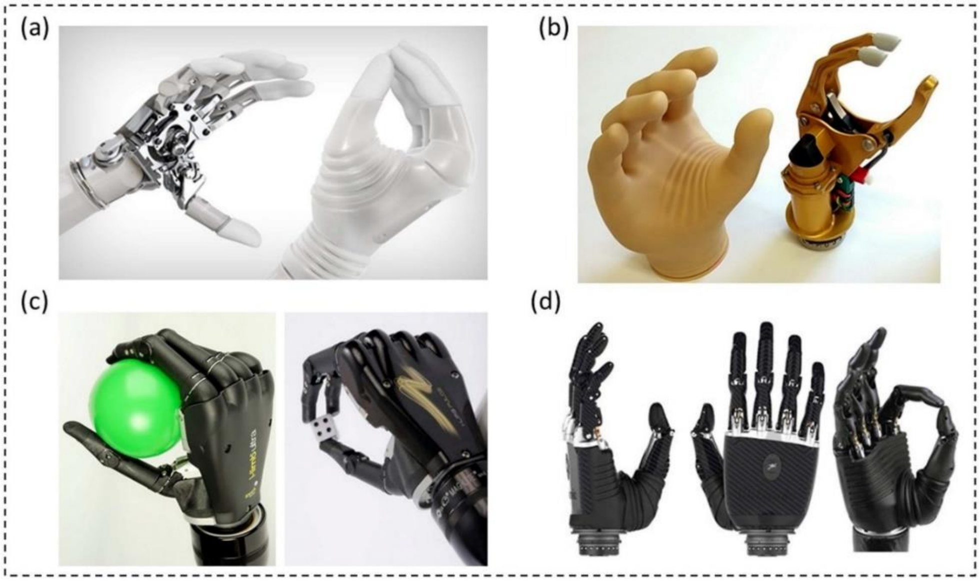

Figure 1.14 Commercially available (a) Michelangelo hand, (b) Ottobock sensor hand, (c) i-Limb quantum hand, (d) Bebionic v3 hand.

The available prosthetic hands with several features are capable of restoring the lost capabilities of amputees. These hands are controlled through single or multi-channel EMG signals and can provide up to eleven grip patterns. However, the main issue with these hands is their excessive cost, which leads to the unavailability of products among amputees, especially in low-income countries.

Parameters	Michelangelo hand	Ottobock sensor hand	i-Limb quantum	Bebionic v3
Weight (g)	420-600	460	474-515	500-590
Size (mm)	180x80x40	190x76x100	180x80x40	200x90x50
DOF	5	1	6	6
Actuators used	5	1	6	6
Actuation method	Dc motor-worm gear	Dc motor-worm gear	Dc motor-worm gear	Dc motor-worm gear
Control scheme	Proportional with pattern recognition	Proportional	Proportional with pattern recognition	Proportional with pattern recognition
Feedback	no	force	no	Vibration
Full closing time (s)	0.37	0.3	0.8	1
Finger movement	Individual as well as combined	combined	Individual as well as combined	Individual as well as combined
Max. grip force (N)	70	100	136	140
No. of grip pattern	7	1	7	11
Battery type	11.1 V, li-ion, 1500 mAh	7.4 V, li-ion, 2200 mAh	7.4 V, li-po, 2400 mAh	7.4 V, li-ion, 2200 mAh
EMG sensor	Two-channel	Single-channel	Multi-channel	Multi-channel
Price	$60,000	$42000	$80,000	$25,000

Table 1.2 Prosthetic hands commercially available in the market.

1.8 Problems

Based on the world health organization (WHO) survey and the international society of prosthetics and orthotics (ISPO), more than 85% of below-elbow amputees residing in developing countries cannot afford to have functional prosthetic devices. Most of these patients are still using a body-powered and cosmetic prosthesis, which cannot fulfill the needs of their daily lives. As per the current survey report and literature review, the chief reasons behind rejection or unavailability of upper-limb prosthesis among amputees are (1) high cost, (2) limited functionality, (3) unnatural control, (4) slow operating speed, (5) complexity (6) lack of dexterity (7) weight and (8) large size.

Amputees residing in developing countries require a simple, affordable, fast, lightweight, robust, and dexterous hand that can perform activities of daily livings (ADLs) with minimum training efforts.

1.9 Research objectives

The foremost aim of this research work is to develop a dexterous, affordable, externally powered prosthetic hand that can fulfill the basic needs of amputees. To achieve this, several sub-objectives have to be accomplished. Therefore, the objectives may be divided into the following parts:

1. Design of low-cost and novel sensors for faithfully detecting the muscular contractions from the residual upper-limb of amputees.
2. Formulation of control strategies for translating the muscle contraction information (using the designed sensor) into control commands for intuitive operation of the prosthetic hand.

3. Development of prosthetic hand, its actuation mechanism, and its socket assembly. Interfacing the microcontroller unit (having control strategy) present within the hand with the designed sensor and power supply.
4. Testing and validation of the designed sensor as well as hand prototype on different amputees.

1.10 Thesis outline

This thesis is broadly divided into seven chapters (see Figure 1.15). Chapter 1 (this chapter) provides a general overview of the amputation and upper-limb prosthesis. This chapter also presents a detailed literature survey and market overview of the externally powered upper-limb prosthesis. The problems for the existing technology and the research objectives of this thesis towards developing affordable, functional transradial prostheses are also described in this chapter.

Chapter 2 describes the design and validation of a wearable surface electromyography (sEMG) sensor for the application of controlling upper-limb prosthesis. The performance of the designed sensor in detecting muscular contractions from the forearm of various subjects was compared with that of a conventional EMG sensor regarding amplitude sensitivity, signal-to-noise ratio (SNR), and response time. Later the designed sensor was successfully tested on amputees for controlling the operation of a custom-made 3D printed prosthetic hand.

The fabrication of a dry electrode-based compact sEMG sensor (an upgraded version of the wearable sEMG sensor designed in chapter 1) is described in chapter 3. The performance of dry electrodes employed in the developed sensor was analyzed with the conventional Ag/AgCl electrodes in terms of electrode-skin impedance and SNR. Moreover, the output performance of the developed sensor was compared with a commercial EMG sensor regarding the signal-to-noise ratio (SNR), sensitivity, and response time. The developed sEMG sensor was further tested on amputees to control the operation of a self-designed 3D printed prosthetic hand.

Chapter 4 presents a multifunctional prosthetic hand that can perform six different grip patterns utilizing a single channel EMG signal from subjects. The EMG signals were acquired from the forearm muscles of various subjects for their six different levels of muscular contraction, using the developed sensor (described in chapter 3). These six levels were classified using the Fuzzy logic system to recognize six predefined hand gestures. The performance parameters such as accuracy, sensitivity, specificity, precision, and F1 score were determined and analyzed to see the effectiveness of classification. Further, the classification based control scheme was implemented in hardware for real-time operation of the self-developed multi-degree-of-freedom prosthetic hand. The hand was able to perform six distinct grip patterns for delicate grasping of various objects utilizing EMG signals from subjects.

Chapter 5 introduces a novel force myography (FMG) sensor for reliably detecting the muscle contractions from the residual upper-limb of amputees to be applied for hand prosthesis control. The static and dynamic characteristics of the designed FMG sensor (i.e., sensitivity, drift, precision, hysteresis, and frequency response) were determined and analyzed using the recorded data to show its effectiveness. The output assessment for simultaneous acquisition of EMG and FMG signals from the flexor muscles of subjects was performed using a two-tailed paired t-test to see the correlation of the designed sensor with the conventional EMG sensor. Further, a successful trial of the FMG sensor was made on five different subjects to control a prosthetic hand in real-time, employing the proportional strategy.

Chapter 6 describes the development of an affordable transradial prosthesis controlled by force myography (FMG) signal as an alternative source to the EMG signal. A hand prototype was specially designed with socket assembly to attach the residual forearm stump of an amputee. A compact-sized FMG sensor was specifically fabricated to attach the residual forearm of amputees for extracting muscle contraction information faithfully. The sensor's performance was authenticated by evaluating its static and dynamic features. Also, its ability to detect muscle activity was compared with that of a standard EMG sensor. The hand prototype and the sensor were integrated with the formulated control strategy to produce a standalone hand prosthesis system. Further, the hand prosthesis was successfully trialed on five transradial amputees for its dexterous operation of grasping objects of daily life.

Finally, the key conclusions and future scope of the thesis are summarized in Chapter 7.

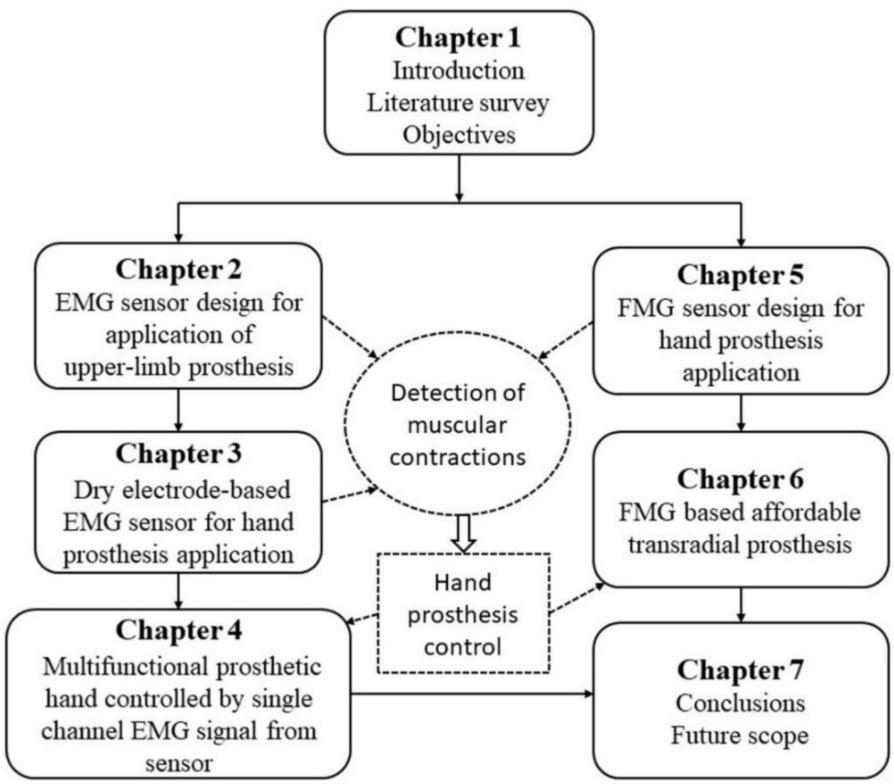

Figure 1.15 Flowchart representation of thesis outline.

1.11 References

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