A Study of Implantable Power Assist and Transhumeral Robotic Prostheses

By

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ABSTRACT

Few decades ago, usually robots were considered as tools in a manufacturing environment to perform welding, spray painting, drilling, material handling or assembling operations. However, recent advances in robotics technology bring a lot of benefit not only in industrial sector, but also in several other sectors. Using the robotics technology in the present world can be found in the manufacturing industry, military application, space exploration, transportation, amusement, rescue operation, household, medical, hazardous environment application etc. In order to acquire unprecedented control and precision, robotics is being introduced to medical application. The continuous advancement of artificial intelligence and robotics technology enables the robots to work in an autonomous manner or in combination with the human in the field of medical applications. Currently robots are used to help or to take care of the patients as a functional substitute of the nurses, as an assistant for the medical surgeon to carry out complex surgery, as a power assist and rehabilitation device for the physically disabled persons, etc. In this thesis, the design and control strategy of implantable power assist and transhumeral robotic prostheses is described which are expected to play a vital role among the physically disabled and above-elbow amputees society, respectively, in near future.

In order to help the physical activities of elderly or physically disabled persons, a new concept of implantable power assist prosthesis is proposed in this thesis that is supposed to assist the human daily life motion from inside of the human body. This thesis presents an implantable 2 degree of freedom (DOF) power assist prosthesis (i.e., inner skeleton robot) that is designed to assist human elbow flexion-extension motion and forearm supination-pronation motion in daily life activities. In this research a prototype of inner skeleton robot is developed that is supposed to assist the motion from inside of the body and act as an actuated artificial joint. The proposed system is controlled based on the activation patterns of the electromyogram (EMG) signals of
the user muscles by applying a fuzzy-neuro control method. A joint actuator with an angular position sensor is designed for the inner skeleton robot and a T-Mechanism is proposed to keep the bone arrangement similar to the normal human articulation after the elbow arthroplasty. The effectiveness of the proposed system has been evaluated through experiments.

The design and control strategy of transhumeral prosthesis for above-elbow amputees is also proposed in this thesis. Transhumeral prosthesis, which is used to compensate for the lost functions of above-elbow (AE) amputees absent arm, is also termed as above-elbow prosthesis or prosthetic arm. Recent progress in biomechatronics technology has facilitated increased mobility of AE amputees in performing daily life activities. However, presently available commercial prosthetic arms have failed to gain wide acceptance among AE amputees due to the discrepancy between their expectations and reality. The main factors causing a lack of interest in presently available prosthetic arms include low functionality and poor controllability. Currently available externally powered AE prosthetic arms provide two or three DOF motions, which are insufficient to generate natural human-like arm motion. In order to improve the quality of life and to increase the mobility of AE amputees in their daily life activities, a 5 DOF AE prosthetic arm is developed in this research work. Control of a multi degree of freedom (DOF) prosthetic arm also remains as a challenging problem. As a result, a new controller strategy for the control of the designed 5 DOF prosthesis is also proposed in this thesis. The proposed prosthesis is supposed to be controlled by using a combination of the electromyogram (EMG) signals and the joint kinematics of the user’s stump arm. A fuzzy rule based controller that uses EMG signals as input information is designed to control the prosthesis elbow and hand motion. Prosthesis forearm and 2 DOF wrist motions are controlled by a task oriented kinematics based controller that uses user’s residual upper arm and prosthesis elbow joint kinematics as input information. An artificial neural network based task classifier is designed for the kinematics based controller that classifies the daily life activities based on the joint kinematics. After the classification, the inverse kinematic technique is applied to calculate the desired wrist and forearm angles for the prosthetic arm in order to realize user’s intended task. The effectiveness of the proposed 5 DOF transhumeral prosthesis has also been evaluated through experiment.
APPROVAL

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CERTIFICATE OF APPROVAL

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DEDICATION

TO MY LOVING PARENTS

APARNA KUNDU AND SUBODH KUNDU
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I would first like to thank Almighty God for the gifts that I have been given that have allowed me to travel so far in life. At last the entire things of this thesis have been completed, which marks the end of a very tough, uneven, but extremely rewarding journey. I would like to express my heartiest gratitude to all who were with me for the achievement of this great success after hard work and commitment.

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CHAPTER 1

INTRODUCTION

Few decades ago, usually robots were considered as tools in a manufacturing environment to perform welding, spray painting, drilling, material handling or assembling operations. Over the last two decades much research has been carried out in the area of robotics and automation technology that broaden the application field of robotics. Recent progress of artificial intelligence and micromachining technology enables the robot to works in an autonomous manner or in cooperation with an operator to carry out complex tasks in diverse environment. Using the robotics technology in the present world can be found in the manufacturing industry, military application, space exploration, transportation, amusement, household, medical application etc. From the field of housekeeping and medical application (like cutting the lawn, vacuum cleaning, deliver items around a hospital, drilling out the bone in the joint replacement surgery etc.) to the field of hazardous application (like within a nuclear reactor, space and underwater application, making the safe of bombs and mines, etc.), the application of robotics technology have realized to the point that near future it is expected to play a vital role in our daily life activities.

1.1 Robotics in Medical Application

In order to acquire unprecedented control and precision, robotics is being introduced to medical application. The application field of medical robotics can be divided into three subfields depending upon the purposes that the robot is served.

1.1.1 Robots for Hospital

A robot could be used in hospitals to help paralytic people, for patient transportation, to take care of postoperative patients etc. Using a master-slave communicating tool, a patient can instruct a robot to bring medicine, adjust the air-
conditioning unit, open the door or turn on the television set. The advancement of the voice recognition technology is also opening the opportunity for a patient to communicate with a robot by voice instruction which in turns relieving nurses from nonprofessional tasks [1].

1.1.2 **Robots for Rehabilitation and Assistive Technology**

The development of robotics technology has already started to prove its significant role in the area of rehabilitation and assistive technology. A robot could be used as an assistive and/or a rehabilitation device for the physically disabled persons [2]. Moreover the progress of the biorobotics and biomechatronics technology provides useful technology for the development of the prostheses and the artificial organs. This thesis is mainly focused on the robots (i.e., robotic prostheses) belongs to this category.

1.1.3 **Surgical Robots for Clinician/Surgeons**

The first generation of surgical robots to carry out minimally invasive surgery is already being installed in a number of operating rooms around the world. The progress of the surgical robots has proved that the former assumption in the surgical world that “a big surgery requires a big incision” is no longer true. Surgical robots offer many advantages in the area of minimally invasive surgery and have made significant contributions to the field in the last twenty years [3].

1.2 **Robotics and Prosthesis**

The term “Prosthesis” can be defined as an artificial replica that replaces human damaged or lost body part. In the field of arthroplasty or joint replacement surgery, prosthesis is defined as an artificial joint that replaces the arthritis affected or damaged human articulation (i.e., bone joint). The primary objective of the arthroplasty is to relieve patients from arthritis pain in addition with restoring joint functions. Though in most of the cases the reason behind the arthroplasty is the arthritis pain, it’s not only the reason. Human articulations can be damaged by severe impacts or unusual stresses. Prostheses are usually made for human main joints such as hip, knee, elbow, shoulder, wrist etc. Long term results of the arthroplasty depend
to a great extent on the quality of prosthesis implantation. The application of the robotics technology in the field of arthroplasty reduces the inaccuracy that occurred in conventional joint replacement surgery such as misalignment, rotation error, resection etc.

On the other hand, in the field of biomechatronics (i.e., the science of fusing artificially intelligent devices with the human body), prosthesis is defined as an artificial substitute for a missing body part. Human limb amputation can causes due to trauma, tumor, congenital, disease, etc. According to biomechatronics, prostheses are considered as those which replace human lost arms and legs. The development of the biomechatronics provides useful technology for the robotic prosthesis. Robotic prosthesis act as an extended body part of the amputee’s by using which amputee’s can be able to perform his/her daily life activities and take care of them by using their own body functions. As a result, robotic prostheses provide an independent life and more productive role of these people in the society.

In addition with the prostheses mentioned above, artificial eyes, teeth, artery, and heart valves are also correctly termed as prostheses. However, the development and control strategy of the two prostheses (i.e., implantable motion assist prosthesis for physically weak people and transhumeral prosthesis for above-elbow amputees) described in this thesis belongs to the first two categories.

1.3 IMPLANTABLE ROBOTIC PROSTHESIS FOR MOTION ASSIST

1.3.1 HUMAN MOTION ASSIST SYSTEM

With each passing year a decrease in birth rate and increase of aging society leads the robotics researchers towards the development of new technology for the elderly persons to help their daily life motion. Few countries of the world have been facing aging problem that is increasing at an alarming rate day by day. Especially in Japan, the number of persons aged 65 and over has reached about 25.31 million in 2005, accounting for 20.1% of the total population. The proportion of the aged population is estimated to be reached about 38.9% in 2050 where as in year 1995, elderly persons made up only 14.6% of the population. In accordance with the current projections, compared to the statistics in 1995, the proportion of elderly population will double to 29.9% in the year 2025. In other words, one of three persons in the population is
estimated to be aged at that time [4]-[6]. Moreover the number of disabled people in
the world today is estimated to be in the region of 500 million [7]. In addition to this,
physical disabilities such as full or partial loss of function in shoulder, elbow or wrist
also appears due to disease processes including trauma, sports injuries, occupational
injuries, spinal cord injuries, and strokes. These levels of disability in combination
with the aging society highlight the necessity for artificially intelligent assistive
devices for the elderly and physically disabled people.

One of the major limits of the human being in performing daily life activities is
their muscle strength. It is clearly evident that muscle strength of those people
mentioned above is less than that of the normal people. In order to alleviate the lack
of muscle strength of these elderly or physically disabled people by using robotic
assistive devices inevitably augment their living style. The idea of assisting human
daily life motion by means of a robotic system is not a new one. In the field of welfare
robotics, much research has been carried out and are ongoing for the physically
disabled people who lost their original body functions in order to support their motion
or to make up their lost function [8]-[30].

1.3.2 EXOSKELETON ROBOT

An exoskeleton robot is usually a wearable motion assist device consisted with
actuators and sensors whose joints correspond to those of the human body [10]-[21].
It is worn by the human and the physical contact between the user and the exoskeleton
allows direct transfer of mechanical power and information signals. In utilizing the
exoskeleton robot, the user provides the control signal for the exoskeleton, while the
exoskeleton actuator provides most of the power necessary for performing the power
assist [10]. Electromyogram (EMG) signals of human muscles are important
biological signals because it reflects the motion intention of a person. The EMG
signal is important for those devices to understand how the human subjects intend to
move. As a result EMG signals have been used as the main input information for the
control of the exoskeleton robot, prosthetic devices and also for the control of robotic
rehabilitation devices [10], [12]-[19], [22], [28]. Moreover, the force caused by the
motion difference between the subject’s limb and the worn robotic exoskeleton is also
used as subordinate input information for the controller [12], [15]. Furthermore, many
researchers choose some form of mechanical control (shoulder switches, cables, etc.)
in lieu of, or in addition with EMG signals to control the motion assist robotic system [24] [31].

The first generation prototype known as Hardyman for the whole body motion assist system was the first attempt to mechanically design a man amplifying exoskeleton [8]. The Hybrid Arm Orthosis (HAO), developed by Benjuya and Kenney at 1990, aimed to provide upper arm motion assistance [9]. That system offered shoulder abduction, elbow flexion and wrist supination motion. At 2001, Jacob Rosen et al. proposed an elbow motion assist exoskeleton robot using muscles EMG signals which has recently expanded to seven degree-of-freedom (DOF) exoskeleton robot for human upper limb motion assist [10], [11]. In order to help physically weak and disabled people in their daily life activities, exoskeleton robots for human elbow, shoulder and forearm motion support have been developing by Kazuo Kiguchi et al. [12]-[15]. The skin surface EMG signals, and the force generated between the exoskeleton robot and the user wrist are fused and used as controller input information for these robots. The sophisticated real time fuzzy neuro control method has been applied to control these robots. Figure 1.1 shows a 6 DOF exoskeleton robot and a 7 DOF exoskeleton robot developed at Saga University, Japan and University of Washington, USA, respectively.

**Figure 1.1:** Human Upper Limb Motion Assist Exoskeleton Robots.

In addition to the upper-limb motion assist exoskeleton robots, several exoskeleton robots have been emerged for lower-limb motion assist during last few years [17]-[19], [24], [25]. Hybrid Assistive Limb (HAL) is a one of the famous
exoskeleton robots which is designed not only to provide motion assist for the elderly persons but also to realize walking aid for the gait disorder people. HAL has the hybrid control system that consists autonomous controller such as posture control and the comfortable power assist controller based on the biological feedback and predictive feedforward[17]-[19]. The Berkley Lower Extremity Exoskeleton (BLEEX) is an another example of lower-limb power assist exoskeleton robot which is basically designed for armed forces to aid soldiers while carrying heavy loads [24]. A one degree of freedom lower-limb exoskeleton called ‘RoboKnee’ has also emerged few years ago which allows the wearer to easily climb stairs and perform deep knee bends while carrying a significant load in backpack [25].

1.3.3 Approach to Implantable Power Assist Robotic Prosthesis

Although the improvement of the exoskeleton robot is continuing day by day, assisting the human limb motion without hindering any other motion or comfortness does not existed yet. The first generation exoskeleton robot “Hardyman” weighted about 3300 kg and the biggest problem was the hydromechanical servo system employed in the leg which did not works properly [12]. Moreover the user had to operate a handle attached to the tip of the exoskeleton to control the robot, which demanded continuous attention of the user. The exoskeleton robots proposed by Kiguchi et al. is made up with wire and pulley mechanisms except the robot designed to provide elbow flexion-extension motion [13] [15]. The proposed mechanism showed the effectiveness for the stationary/wheel chair operated users. However, for daily life activities or in other words for mobile user’s the effectiveness and/or comfortness is yet to be proved. The anthropometric seven DOF powered exoskeleton for upper limb proposed by J. Rosen and J. C. Perry also comprises the cable driven mechanism [32]. The proposed design demands a large stationary base to locate the four motors. The torque from the motors is transferred to the corresponding joint by pulley-cable mechanism. So, the question of comfortness or convenience is still arises in this case also.

Usually exoskeleton robot is designed to attach on the lateral side of the human limb that require user’s attention to avoid interaction with the surrounding environment. The exoskeleton robot is not convenient in certain daily life activities such as during taking shower or sleeping. Moreover, as the center of rotation of the
exoskeleton robot and that of the human limb are not the same, sometimes, it is very
difficult to obtain the full range of limb motion with desired degree of freedom by
using the exoskeleton robot. Furthermore, the exoskeleton robot used surface
electromyogram signals as controller input information. Surface EMG signals are
easier to measure than intramuscular EMG signals as they don’t require any
implantation of electrode inside of the muscles. However, sometimes a surface
electrode picks up the EMG activity from the all active muscles in its vicinity, while
the intramuscular EMG is highly selective. Moreover, the combination of muscle
tissue, adipose (fat) tissue, skin and the skin electrode interface behaves like a
nonlinear low pass filter which attenuates and distorts the intramuscular EMG signals
[33]. Therefore, estimation of human motion intention using the surface EMG signals
is much more difficult than using the intramuscular EMG signals. Moreover, it is
sometimes very difficult to pick up the EMG signals from some deep muscles using
surface electrodes as there is overlapping of several muscles in a limited space inside
of the human arm. Brachialis muscles can be considered as an example for this case.
Brachialis is referred as a major flexor part of the elbow, but due to the fact that it is
almost impossible to measure the EMG signal of brachialis by noninvasive technique
[10] [34].

To alleviate the inconveniency possessed by the exoskeleton robot and the surface
EMG signal as a controller input information for the exoskeleton robot, and to keep
the physical activities of the elderly person with proper amount of limb motion with
desired DOF, a new concept of implantable robotic power-assist prosthesis is
proposed in this thesis. Since the proposed prosthesis is expected to assist the power
from inside of the human body, it is termed as “Inner Skeleton Robot”. The principal
difference between the exoskeleton and the inner skeleton power assist system is
shown in Fig. 1.2. A prototype of an implantable 2 DOF inner skeleton robot that has
designed to assist human elbow flexion-extension motion and forearm supination-
pronation motion is proposed in this thesis. The proposed inner skeleton robot is
expected to be implanted inside of the human arm by elbow arthroplasty and acts like
an artificial joint. Figure 1.3 shows an example of artificial elbow joint i.e., elbow
prosthesis [35].

Artificial elbow joint or elbow prosthesis is used in joint replacement surgery or
elbow arthroplasty to get relief patients from arthritis pain in addition with restoring
joint function and stability [36]-[38]. The intermediate part of the elbow
Implantable Prosthesis/ Inner Skeleton Robot – Assist Motion from Inside of the Body
Exoskeleton Robot – Assist Motion from Outside of the Body

**FIGURE 1.2**: Comparison of Exoskeleton and Inner Skeleton Motion Assist System.

**FIGURE 1.3**: The GSB III Elbow Prosthesis [35].

prosthesis that adjoins the humeral and ulnar side acts like a hinge joint. The objective of this research work is to develop a prototype of inner skeleton robot that is supposed to substitute the existing elbow prosthesis by incorporating an actuator at the hinge joint of the elbow prosthesis or in other words developing an actuator that can serves as an elbow prosthesis and assist the joint motion from inside of the body in addition with meet the basic criteria fulfilled by the elbow prosthesis. Moreover, as the human elbow complex provides not only the elbow flexion-extension motion but also the forearm supination-pronation motion, the proposed inner skeleton robot has designed to provide 2 DOF motion. It is obvious that the robotic power assist elbow prosthesis will be more attractive for the persons seeking to get elbow arthroplasty. The inner skeleton robot for elbow joint (i.e., implantable power assist elbow prosthesis for elbow arthroplasty), however, still possesses challenging problems as a substitute for normal elbow articulation.

In order to realize the natural motion assist from inside of the body, the approach is to design the inner skeleton robot that can imitate the functions provided by the normal human articulation. Human elbow complex provides two DOF with specific
range of motion and the specific shape of the bony surfaces which form the articulations, keep the movement within the normal range. The implantable power assist system can be realized by replacing the bony surfaces of an articulation with the stator and rotor of an implantable joint actuator that is designed based on the shape and range of motion of that articulation. Although there are many available actuation technologies used in HRI (Human Robot Interaction) robot design [39], inner skeleton motion assist is still difficult to accomplish by using a conventional actuator. Numerous actuation technologies especially traditional rotary electric motor is used in upper limb power assist exoskeleton robot, prosthetic arm and biologically inspired humanoid robot arm to generate human-like specific range of arm motions. In recent years, a number of actuators capable of generating multi DOF motion have also been proposed for robot manipulators [40]-[42]. However, the implantable joint actuator has not existed yet. In this study, a prototype of implantable joint actuator, an angular position sensor, and a T-mechanism have been developed to make the proposed inner skeleton robot (i.e., implantable power assist prosthesis). The principle mechanism of the joint actuator is similar to that of the switched reluctance actuator. The range of motion and, the stator and rotor of the actuator have been designed according to the human joint motion. Usually, actuators for hinge motions are constructed from DC motors and reduction gears which require high cost and maintenance [43]. Traditional DC motor consists of complex mechanical structure and includes permanent magnet or windings on the rotor. The designed joint actuator provides simple structure and complete absence of permanent magnet or windings on the rotor makes it inexpensive and suitable to implant inside of the body. The piezoelectric ultrasonic actuators are also claimed to be lighter, more compact and larger generating torque than the conventional electromagnetic actuators, but they could not gain wide acceptance due to their high cost and the necessity of high amount of voltage at ultrasound frequencies. Furthermore, the movable part of the ultrasound actuator becomes locked when the power is turned off. The direction of rotation of the designed actuator can be reversed by changing the stator excitation sequence which is relatively simpler than the conventional DC motor that leads to use relatively inexpensive, small size semiconductor to control the actuator instead of using relatively expensive, larger size motor driver. Since the joint actuator does not provide complete rotary motion and, its stator and rotor are supposed to attach with the bony surface of the articulation, it is inconvenient to use a conventional position sensor to determine the rotor position for
angular feedback to control the actuator. As a result, a contact type potentiometric angular position sensor has been designed for the semicircular joint actuator. Although the non-contact type position sensor is more attractive, but they are more sensitive and susceptible in a changeable magnetic field than the contact type potentiometric angular position sensor. The actuator and the position sensor can be designed for both of the elbow flexion-extension motion and forearm supination-pronation motion. However, for the prototype robot described in this thesis, the proposed actuator with the position sensor has designed only for elbow flexion-extension motion and a DC motor has used for forearm supination-pronation motion. The forearm supination and pronation motions has the uttermost importance to the human being as it, in combination with the upper arm motions control the position and rotation of the hand and allow us to perform daily activities such as feeding oneself, drinking a glass of water, turning a key in a lock, performing personal hygiene, etc. [44] [45]. In normal human elbow complex, during supination motion the radius and ulna are parallel to each other and during pronation motion the radius bone of the forearm rotate and crosses over the ulna bone [34] [46]. To keep the similar bone arrangement after the elbow arthroplasty, we introduce a T-mechanism for the proposed system. As discussed earlier, the motion intension of a person can be monitored from the muscles EMG signals. Therefore, the EMG signals can be used as the control input signals for the proposed system in order to realize the natural motion assist. It is expected to assist the motion from inside of the human body and uses the intramuscular EMG signals as the controller input information. Since the estimation of motion intension using the intramuscular EMG signal is much easier than the surface EMG signals, better control performance can be expected by the proposed inner skeleton robot than the exoskeleton robot. A sophisticated fuzzy-neuro control method, in which the effect of a muscle common to both of the elbow and forearm motion is considered, has been used to control the proposed inner skeleton robot that enables the cooperative motion of the human elbow and forearm motion [15]. Since the role of each muscle is changed according to the arm posture, the magnitude of the EMG signals is also affected by the arm posture. In order to cope with this problem, multiple fuzzy-neuro controller have been applied for the control of the exoskeleton robot and the similar approach is also implement here [47]. The proposed inner skeleton robot can be expected to be an artificial joint for future generation.
1.4 TRANSHUMERAL PROSTHESIS FOR ABOVE-ELBOW AMPUTEES

1.4.1 TRANSHUMERAL PROSTHESIS

“Transhumeral Prosthesis” can be defined as a prosthesis that is designed for the people who have lost their arm just above the elbow joint (i.e., above-elbow amputees). Human arm amputation can caused due to congenital (birth defect), tumor, trauma, disease, etc. Circulatory disease, cancer and infections are considered as the major categories of disease which may require surgical removal of human arm. Moreover, the civil wars and more specifically wars in Iraq and Afghanistan producing an unprecedented number of amputees. Although nothing can ever become a perfect substitute for a missing arm, the intension of the transhumeral prosthesis is to compensate for the lost functions of the above-elbow (AE) amputees absent arm, so that they can lead an independent life and play more productive role in the society.

1.4.2 HISTORICAL EVOLUTION OF THE PROSTHETIC ARM

Prostheses have been found around for thousands of years, however real advancement and fabrication of the prostheses have started about 500 years ago [48]. According to the medical museum exhibited at the University of Iowa titled “History of Prostheses”, earliest prostheses were used by soldiers dating back to 484 B.C. Hegesistratus; a Persian soldier around 490 B.C. cut off part of his own feet in order to escape from the prison and later replaced it with a wooden foot [49]. In 61 A.D., Pliny the Elder wrote about the Roman General Marcus Sergius who had lost his right arm during the Second Punic War (218-201 B.C.). Later he had replaced that by an iron arm to support his shield and he returned to battle [50]. During the middle ages, 15th and 16th centuries cosmetic prostheses were usually made from iron. At that period, blacksmiths and armor makers designed the prostheses for the soldiers after modeling their suits of armor. In the 16th century, the great French arm surgeon Ambrose Pare, designed several limb prostheses in addition with practicing surgical amputation. In 1818, Peter Baliff appears to have been the first person to introduce the use of the trunk and shoulder girdle muscles as sources of power to move the prosthetic arm. In 1844, the first transhumeral amputation replacement used Baliff’s principle to apply flexion for the elbow joint [50]. The prosthetic arm using this
concept is termed as “Mechanical” or “Body Powered” prosthetic arm and still extremely famous among the amputees society. By 1860, the Crimean and Italian campaigns of the French Empire left many soldiers in need of prostheses, and their call was answered by the Comte de Beaufort [48] [50]. The Comte de Beaufort designed several limb prostheses using the articles of clothing, pulleys and levers. After the World War I and II, a tremendous loss of manpower in USA and Europe served as a catalyst for the rapid development of the prosthetic arm. In 1948, N. Wiener proposed the concept of Cybernetics i.e., the study of control and communication between the human and the machine [51], which plays an important role later for the development of the prosthetic arm. In 1949, Samuel Anderson created the first electrically powered prosthetic arm using the external power with support from the US Govt. and IBM. The first myoelectric arm was developed by Russians in 1958 and later on Otto Bock Company revealed the commercially available prosthetic arm for general application which was the first made versions of the Russian design [50].

1.4.3 **CLASSIFICATION OF THE PROSTHETIC ARM**

Prosthetic arms can be grouped into three general categories:

1) **Non-functional or Cosmetic Prosthetic Arm** – As the name implies functioning of these prostheses has less priority than the appearance, weight, wearing comfort and easy handling. These are the oldest and available for 2000 years. Though cosmetic prostheses offer a more natural look and feel, they sacrifice functionality and versatility while also being relatively expensive [52].

2) **Mechanical or Body Powered Prosthetic Arm** – The power to operate these prostheses comes from the user’s own body. In this system, the user wears a harness that translates the shoulder motion into elbow flexion motion and action of gravity force generates the elbow extension motion. The earliest model of this prosthetic arm was the Ballif arm [53]. These prosthetic arms are light weight and less expensive than the others however it requires large amount of forces to actually move the elbow [54].
3) Externally Powered Prosthetic Arm – Most advanced commercially available prosthetic arm in which power to operate the prosthetic arm comes from the external sources such as electric motor and battery pack. Most of these prostheses are operated by using user’s stump arm muscles EMG signals. This type of prostheses provides greater proximal functions, increased cosmetic appeal but also tend to be much heavier and expensive than any of the other categories [54]. The proposed transhumeral (i.e., above-elbow) prosthetic arm described in this thesis belongs to this category.

1.4.4 Present State and Proposed Transhumeral Prosthesis

Recent progress in biomechatronics technology brings a lot of benefit to increase the mobility of above-elbow (AE) amputees in their daily life activities. A transhumeral or AE prosthetic arm is used to compensate for the lost functions of the AE amputees absent arm. A number of commercial prosthetic arms have been developed since last few decades. However, many amputees have not used them due to the discrepancy between their expectations and the reality. The main factors causing a loss of interest in presently available prosthetic arms include low functionality and poor controllability [55].

Since the concept of Cybernetics proposed by N. Wiener [51], a number of research works have already been carried out and are ongoing for the development of prosthetic arm. At present, Utah arm, Boston Elbow, and Otto Bock are considered as the pioneers in this field which are shown in Fig. 1.4 [56]-[59]. However, currently, commercial prosthesis available on the market for the AE amputees provides a limited DOF. Most of these prostheses provide elbow flexion-extension motion with a terminal device attached at the end. In addition to the elbow motion, some prostheses provide forearm supination-pronation motion and a single DOF at the terminal device for grasping object. Some passive DOF, which are useful to generate an optimal pre-determined configuration during performing certain tasks [60], are sometimes included in the prostheses. Commercially available expensive cosmetic prostheses offer a more natural appearance and simple control. However, their dexterity is relatively very poor compared to the human arm. Human arm generates precise and complex motions during daily life activities which are almost impossible to be
generated by using a limited DOF prosthetic arm. As a result, the presently available commercial prostheses have failed to gain wide acceptance among AE amputees.

![Utah Arm] ![Boston Elbow] ![Otto Bock Arm]

**FIGURE 1.4**: Commercially Available Externally Powered AE Prosthetic Arm [61].

In order to improve the quality of life of AE amputees and to increase their mobility in daily life activities (like, eating, drinking, dressing, brushing etc.), a 5 DOF externally powered transhumeral prosthesis is proposed in this thesis. The prosthesis is designed to generate elbow flexion-extension, forearm supination-pronation, wrist flexion-extension and radial-ulnar deviation, and hand cylindrical grasp-release motion. Currently, no commercial transhumeral prosthesis provides a combination of wrist flexion-extension and radial-ulnar deviation motion, which have utmost importance to perform daily life activities. In recent years, a number of prostheses capable of generating multi-DOF motion have been proposed for upper limb amputees [60], [62]-[66]. However, none of these provide a combination of forearm and 2 DOF wrist motion with the exception of an arm designed for above-wrist amputees to provide wrist flexion-extension and forearm motion [65].

To imitate human arm function during prosthesis design, it is essential to use flexible actuators that can act like human muscles. However, electroactive polymer, Mckibben muscles, and shape memory alloy wires have not developed enough yet to use as an actuator for the prosthetic devices. In this study, traditional rotary DC electric motors are used to generate the required motion. In the human arm, muscles acting as actuators for elbow motion carry the load of the forearm and hand during elbow movement. The forearm consist the muscles that generate forearm, wrist and hand grasp-release motion. However, each of these muscles does not carry the load of others during the participation in corresponding joint motion. A similar principle is applied to place the actuators (i.e., motors) in the proposed prosthesis to mimic human anatomy. Moreover, placing the hand motor in the forearm part not only reduces the inertia effect but also provides sufficient grasping area in the palm.
In commercial and previously proposed prostheses, a terminal device is mounted on a circular rotating unit to provide forearm supination-pronation motion [61], [63]-[65], which is different from their biological counterpart. As mentioned earlier, in the human arm, the radius and ulna bone of the forearm are almost parallel to each other in supinated forearm position and the radius bone crosses over the ulna bone in pronated position. In order to mimic human-like forearm motion, two shafts are used as radius and ulna bone in the forearm and a T-mechanism is used that keeps the shaft arrangement like radius and ulna bone during forearm supination-pronation motion. Moreover, these two shafts connect the forearm part with the wrist part like the human arm. A ball joint and wire tension mechanism is used in the wrist part to accomplish two DOF wrist motions. The pulley and Bowden cable mechanism are used to generate wrist and hand motion. The designed prosthesis provides sufficient joint torque compare to the normal human arm, and weight and length of the prosthesis are kept almost similar to the normal human arm.

At present, the most advanced commercially available externally powered transhumeral prostheses are myoelectric prostheses. The skin surface electromyogram (EMG) signals of amputee’s stump or residual muscles are used as input signals to control the myoelectric prosthesis [48], [56], [58], [62]-[73]. The EMG signals are among the most important biological signals that directly reflect human motion intention. The EMG signals are measured with surface electrodes which are then amplified and filtered properly to extract the feature of the EMG signals. The extracted feature values of the EMG signals are used to control the electromechanical active joints of the prosthesis and actuate the prosthesis arm segments. However, it is not easy to properly control the multi-DOF prosthetic arm for AE amputees with a limited number of EMG signals. For an AE prosthetic arm, it is difficult to generate the forearm, wrist and hand motions by using only the limited number of EMG signals from remaining arm muscles. As a result, most AE prosthetic arms provide single or 2 DOF motions.

Recently, a Targeted Muscle Reinnervation technique was shown to improve the myoelectric prosthesis control for higher level arm amputees [48], [74]. In this technique, the active nerves from the residual part of the arm are transferred to the muscles of another part of the body. The EMG signals of those muscles are then used to control the prosthetic arm. However, this is still under clinical trial and the amputees have to undergo a surgical operation to implement this technique for
prosthesis control. A concept for controlling a prosthetic arm using a muscle tunnel cineplasty was proposed a few decades ago [75]. In this technique, a part of stump muscles are fitted with the part of the prosthetic arm to control the active joint of the prosthesis. This procedure to control an externally powered prosthetic arm also requires surgical procedures in conjunction with a physiologically appropriate prosthesis controller [76]. Moreover, this procedure is not suitable for the control of a multi-DOF prosthetic arm due to the limited number of remaining muscles and complex surgical procedure. The control of a prosthetic arm by using the kinematic data of the remaining fully functional normal arm was also proposed [77]. However, this technique is suitable to carry out only those tasks that require cooperation of both arms. Another approach for control of a prosthetic arm is the extended physiological proprioception (EPP) technique that uses the residual arm joint kinematics as controller input information [78]-[79]. A control method for a 2 DOF prosthetic arm using only shoulder flexion-extension angle was proposed [80]. According to this controller technique, an amputee has to operate an electromechanical device as a linkage switch by the contralateral hand to choose an appropriate linkage between the input and output. Consequently, it is not very useful for daily life activities, as it demands the continuous attention of the amputee. Estimating the elbow and forearm motion from shoulder flexion-extension and abduction-adduction angle by using a pattern recognition system was also proposed [81]. However, this system considered only three activities of daily living. For the control of a multi-DOF prosthetic arm, it is also necessary to provide control input signals to generate 2 DOF wrist motions, which are not considered in the previous works noted above.

In this thesis, a controller (i.e., a combination of the muscles EMG signals based controller and the task oriented kinematics based controller) is also proposed for a 5 DOF prosthetic arm. The proposed controller is designed based upon the assumption that the biceps and triceps muscles are present in the stump arm and can be activated by the amputee’s intension. A fuzzy rule based EMG controller that uses the EMG signals of biceps and triceps muscles as input information is designed to control the prosthesis elbow and hand motion. The ultimate intention of the proposed prosthetic arm is to provide an independent life to amputees so that they can perform regular daily activities by using their prosthesis. Activities that are essential and frequently performed in daily living are considered in this study. In order to control the prosthetic arm wrist and forearm motions, the amputee’s intended activity is identified
in the first step. A multilayer artificial neural network that classifies the daily life activities using the amputee’s shoulder and prosthesis elbow kinematics is designed. For the classified activity, the desired trajectory of the hand with respect to a fixed shoulder coordinate system is estimated based on the nature of the task in the second step. In the third step, the inverse kinematic technique is applied to calculate the desired 2 DOF wrist and forearm motions. Experiments have been carried out using motion capture data and EMG signals to evaluate the effectiveness of the proposed controller.

1.5 Contributions Made by This Thesis

The contributions of the research works described in this thesis can be outlined as follows:

- Proposed a new concept of inner skeleton motion assist system for elderly and physically disabled people.
- Developed a prototype of 2 DOF implantable power assist prosthesis (i.e., inner skeleton robot) for human elbow and forearm motion assist. A prototype implantable joint actuator, an angular position sensor, and a T-mechanism have been developed to make the proposed inner skeleton robot.
- Proposed the mechanical design and controller strategy for a 5 DOF transhumeral (i.e., above-elbow) prosthetic arm for above-elbow amputees to increase their mobility in daily life activities.
- Since the existing above-elbow prosthetic arms have failed to gain wide acceptance among amputees society due to their low functionality and poor controllability, a 5 DOF prosthetic arm is developed that is expected to improve the quality of life of above-elbow amputees.

1.6 Thesis Outline

This thesis describes the fundamental concept, mechanical design and controller approach of a 2 DOF implantable power assist and a 5 DOF transhumeral prosthesis for physically disabled and above-elbow amputees, respectively. Plausibly these have been arranged into several chapters for handiness.
Chapter 2: Human Arm Motion – Toward the Prosthesis Design

This chapter briefly describes human arm anatomy as the intension of this research work is to develop prosthesis for the human arm. The motions provided by the human shoulder joint, elbow complex, wrist and hand, and their range of motions are described here. Moreover, muscles which are responsible for elbow and forearm motion are outlined here.

Chapter 3: Arthroplasty and Power Assist Prosthesis

An introductory description of the conventional elbow prostheses and elbow arthroplasty is described here. Later on, the detailed description about the design and control strategy of the proposed implantable power assist elbow prosthesis is provided here. The experimental results provided here show the effectiveness of the proposed concept.

Chapter 4: Design of the Transhumeral Prosthesis

The development of a 5 DOF above-elbow prosthetic arm is described here. The mechanical design of the prosthesis elbow, forearm, wrist and hand joint are depicted here. The actuation system of the prosthetic arm and the range of each active joint are explained here. The experimental results show the controllability and range of motion of the prosthetic arm.

Chapter 5: Control Strategy of the Transhumeral Prosthesis

The control approach for the designed transhumeral prosthesis is elaborately described here. The experimental results depicted here ensured the effectiveness of the proposed controller strategy.

Chapter 6: Conclusions and Future Works

This chapter includes the summary and concluding remarks about all findings of the research works presented in this thesis together with some future recommendations.
CHAPTER 2

HUMAN ARM MOTION – TOWARD THE PROSTHESIS DESIGN

A kinematic pair can be defined as a set of adjacent links connected with one joint. The assembly of several kinematic pairs is referred to as a kinematic chain. If each link in a kinematic chain is a part of no more than two pairs, it is defined as a serial kinematic chain. Moreover, if one end of the chain is free to move it can be referred to as an open type kinematic chain. As a result, human arm can usually be considered as an open type serial kinematic chain since its distal segment is free to move. The human arm is mainly composed of three chained mechanism, shoulder complex, elbow complex and wrist. This chapter summarized about motions of the arm provided by its main joints and about muscles that generate these motions.

2.1 SHOULDER COMPLEX

The complex construction of muscles and bones in the shoulder provides a unique mobility and is considered as the most mobile joint in the body. The human shoulder complex is made up of three bones, the clavicle, scapula, and humerus and consists of four articulations, glenohumeral, claviculoscapular, sternoclavicular, and scapulothoracic. The human shoulder complex is shown in Fig. 2.1.

The generic term “Shoulder Joint” is commonly refers to the glenohumeral joint which is a synovial ball-and-socket joint. It is formed by the proximal part of the humerus (humerous head) and the female part of the scapula (glenoid cavity). The claviculoscapular joint which is also known as acromioclavicular joint is formed by the lateral end of the clavicle and the acromion of the scapula. The sternoclavicular joint is a compound joint which has two compartments separated by articular disks. It is formed by the parts of clavicle, sternum, and cartilage of the first rib. In true sense, the scapulothoracic joint can not be considered as a joint as it is a bone-muscle-bone...
articulation which is not synovial. It is formed by the female surface of the scapula and the male surface of the thorax. However, it is considered as a joint when describing motion of the scapular over the thorax [34] [46].

**FIGURE 2.1:** The Shoulder Complex.

The shoulder girdle which is commonly moves as a unit is made up by the scapula and clavicle bone. The shoulder complex can be viewed as a complex mechanism comprising two individual mechanisms: (a) the shoulder girdle, in which the sternum with the rib cage is considered as the frame and clavicle, and scapula as the moving links and (b) the humerus as a moving link and the scapula, and clavicle as a frame [46]. When considering each of the four articulations separately, one would expect the shoulder complex provides 12 DOF as each of the articulations has 3 DOF. However, since the scapula and the clavicle is move conjointly (i.e., shoulder girdle), the shoulder complex provides 7 DOF for the human arm movement. The shoulder girdle provides the 4 DOF and the remaining 3 DOF motion is provided by the glenohumeral joint. For ease of explanation of these motions, human body sagittal, coronal and transverse planes are shown in Fig. 2.2.

The kinematics of the shoulder motions provided by the glenohumeral joint are used as a part of controller input information for the control of the 5 DOF transhumeral prosthetic arm. The human glenohumeral joint provides shoulder flexion-extension, abduction-adduction and internal-external rotation motions, which are shown in Fig. 2.3 (a), (b), and (c), respectively. Shoulder flexion-extension motion is occurred along the sagittal plane. During flexion motion, angle between the articulating elements is increased whereas during extension motion angle is decreased. Shoulder abduction-adduction motion takes place along the coronal plane.
In shoulder abduction motion, the human arm swings away from the sagittal plane and the reverse movement is adduction. In shoulder internal rotation (i.e., medial rotation), the anterior surface of the arm is rotate inward, toward the ventral surface of the body. And the opposite movement is the shoulder external rotation (i.e., lateral rotation). Usually, the limitation of the movable range of human shoulder are 180° in flexion, 60° in extension, 180° in abduction, 75° in adduction, and about 90° for both of the internal and external rotation (at 90° flexed elbow position) [14] [82].

**Figure 2.2**: Human Body Planes.

**Figure 2.3**: The Human Shoulder (a) Flexion – Extension (b) Abduction-Adduction and (c) Internal – External Rotation Motion.
2.2 Human Elbow Complex

Human elbow complex which is shown in Fig. 2.4 is formed by three bones, the humerus of the upper arm and the radius, and the ulna of the forearm, and includes elbow joint and radioulnar joint. The elbow joint is a complex hinge joint that permits elbow flexion-extension motion and involving the humeroradial joint, between humerus and radius, and the humeroulnar joint, between humerus and ulna. The largest and strongest articulation at the elbow is the humeroulnar joint, where the trochlea of the humerus projects into the trochlear notch of the ulna. Much of the stability comes from the interlocking of these two bones. The humeroulnar joint is a hinge joint. The other portion of the elbow joint consists of the humeroradial joint formed by the capitulum of the humerus and the flat superior surface of the head of the radius. The humeroradial joint is a ball-and-socket joint; however its close association with humeroulnar and radioulnar joint restricts the joint motion from 3 to 2 DOF. In the elbow joint, the humerus is the male member. The radioulnar joint is formed between radius and ulna, and permits forearm supination-pronation motion. The elbow joint, when considered as an entirely, is a hinge joint; and radioulnar joints are pivot joints with 1 DOF. As a whole, human elbow complex allows two DOF, elbow flexion-extension motion and forearm supination-pronation motion which are

FIGURE 2.4: Human Elbow Complex.

FIGURE 2.5: Human Elbow and Forearm Motion.
shown in Fig. 2.5(a) and (b), respectively. Elbow flexion-extension motion is occurred along the sagittal plane. During flexion motion the angle between the forearm and the upper arm is increased whereas during extension motion the angle is decreased. Forearm supination-pronation motion is happened along the transverse plane. Pronation is the motion when the human turns his/her palm down so that he/she can able to see the backward face of the palm and supination is the motion when the human turns his/her palm up. During pronation motion, the radius bone rotates and crosses over the ulna bone, whereas during supination motion they stay just about parallel in position. The radius and the ulna bone position during supination-pronation motion are shown in Fig. 2.6. When the arm is completely extended, human forearm supination-pronation motion is conjoint with the shoulder internal-external rotation motion. Usually, the limitation of the movable range of elbow flexion-extension motion is 145 degrees in flexion and 5 degrees in extension and that of the forearm supination-pronation motion is 80-90 degrees in supination and 50-80 degrees in pronation. The combination of elbow flexion-extension and forearm supination-pronation motion, which are generated by the elbow complex, is extremely essential for the accuracy of various minute movements of the hand.

![Humerus, Radius, Ulna](image1.png)

**FIGURE 2.6:** Radius and Ulna Bone during Forearm Supination-Pronation Motion.

### 2.3 The Wrist

The wrist, or carpus, is a deformable anatomic entity formed by eight carpal bones that connects the forearm with the hand. The carpal bones exist in the wrist at two rows, four proximal carpal bones and four distal carpal bones which are shown in Fig. 2.7. The wrist contains several joints, including radiocarpal joint, several intercarpal joints, and five carpometacarpal joints [46]. The generic term “wrist joint” is usually referred to radiocarpal joint which is formed between the distal end of the
radius and the proximal row of the carpal bones (except the pisiform bone). When taken as an entity, the wrist joints are considered one joint, called the wrist joint and permits two DOF, wrist flexion-extension motion and radial-ulnar deviation motion. Wrist radial-ulnar deviation motion is also named as wrist abduction-adduction motion. During wrist flexion, the palm approaches to the anterior surface of the forearm and the reverse movement is the extension. Wrist radial deviation is bending the wrist toward the thumb side and the opposite movement is the ulnar deviation. Human wrist flexion-extension and radial-ulnar deviation motions are shown in Fig. 2.8(a) and (b), respectively. In the human wrist, 2 DOF motions are generated with an instantaneous center of rotation, although the path of centrode is small [83]. However, customarily, the path of the center of rotation is ignored and the rotation axes for flexion-extension and radial-ulnar deviation are considered as a fixed one [46]. Usually, the movable range of wrist motion is 65°-85° of flexion, 50°-70° of extension, 15°-25° of radial deviation, and 25°-45° of ulnar deviation [84] [85]. The movable ranges of human shoulder, elbow and wrist motion are listed in Table 2.1.

**FIGURE 2.7:** Human Wrist – Right Hand, Anterior View.

**FIGURE 2.8:** Wrist (a) Flexion-Extension and (b) Radial-Ulnar Deviation Motion.
### Table 2.1: Human Shoulder, Elbow and Wrist Range of Motion (ROM)

<table>
<thead>
<tr>
<th>Articulation Motion</th>
<th>ROM (in Degree)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shoulder Flexion-Extension</td>
<td>180° - Flexion &amp; 60° - Extension</td>
</tr>
<tr>
<td>Shoulder Abduction-Adduction</td>
<td>180° - Abduction &amp; 75° - Adduction</td>
</tr>
<tr>
<td>Shoulder Internal-External Rotation</td>
<td>90° for Both of Internal &amp; External Rotation</td>
</tr>
<tr>
<td>Elbow Flexion-Extension</td>
<td>145° - Flexion &amp; 5° - Extension</td>
</tr>
<tr>
<td>Forearm Supination-Pronation</td>
<td>80°-90° - Supination &amp; 50°-80° - Pronation</td>
</tr>
<tr>
<td>Wrist Flexion-Extension</td>
<td>65°-85° - Flexion &amp; 50°-70° - Extension</td>
</tr>
<tr>
<td>Wrist Radial-Ulnar Deviation</td>
<td>15°-25° of Radial deviation &amp; 25°-45° of Ulnar deviation</td>
</tr>
</tbody>
</table>

Human hand possesses 22 DOF and is capable of realizing complex and precise daily life activities. Although, natural human hand provides several grasp (spherical, cylindrical, lateral, tin pinch, etc.) and gesture abilities, in this thesis, only cylindrical grasping is considered for the proposed 5 DOF transhumeral prosthesis. Human hand cylindrical grasping as shown in Fig. 2.9, is important while grasping a can, a bottle, drinking a glass of water etc.

![Figure 2.9: Human Hand Cylindrical Grasp-Release Motion.](image)

### 2.4 Muscles That Move the Arm

The human arm consists of a number of muscles which work together in groups contracting and relaxing to generate the fine movement of the arm to perform various daily life activities. Human upper limb consists not less than 21 muscles, which can be even divide in several bundles attached on several bones [34] [87]. These muscles can be divided into several groups depending on the bones they move and the DOF they control. Muscle electromyogram (EMG) signal is one of the most important biological signals as it reflects the motion intention of a person. These muscles EMG signals are used to control the robotic prostheses or motion assist devices. The research work described in this thesis used EMG signals of those muscles which are
responsible for the motions generated at the elbow complex. As a result, only those muscles which generate the motions at the elbow complex are discussed here.

**Table 2.2: Muscles for Elbow Complex**

<table>
<thead>
<tr>
<th>Action at Elbow</th>
<th>Muscle</th>
<th>Origin</th>
<th>Insertion</th>
</tr>
</thead>
<tbody>
<tr>
<td>Flexion</td>
<td>Short Head of Biceps</td>
<td>Originate on the Scapula</td>
<td>Radial Tuberosity</td>
</tr>
<tr>
<td></td>
<td>Long Head of Biceps</td>
<td>Originate on the Scapula</td>
<td>Ulnar Tuberosity</td>
</tr>
<tr>
<td></td>
<td>Pronator Teres</td>
<td>Medial Epicondyle of Humerus and Coronoid Process of Ulna</td>
<td>Distal Lateral Surface of Radius</td>
</tr>
<tr>
<td>Extension</td>
<td>Triceps Long Head</td>
<td>Originate on the Scapula</td>
<td>Olecranon of Ulna</td>
</tr>
<tr>
<td></td>
<td>Triceps Lateral Head</td>
<td>Originate on the Scapula</td>
<td>Olecranon of Ulna</td>
</tr>
<tr>
<td>Supination</td>
<td>Extensor Carpi Radialis Longus</td>
<td>Lateral Supracondylar Ridge of Humerus</td>
<td>Base of 2nd Metacarpal Bone</td>
</tr>
<tr>
<td></td>
<td>Short Head of Biceps</td>
<td>Originate on the Scapula</td>
<td>Radial Tuberosity</td>
</tr>
<tr>
<td></td>
<td>Long Head of Biceps</td>
<td>Originate on the Scapula</td>
<td>Ulnar Tuberosity</td>
</tr>
<tr>
<td>Pronation</td>
<td>Pronator Teres</td>
<td>Medial Epicondyle of Humerus and Coronoid Process of Ulna</td>
<td>Distal Lateral Surface of Radius</td>
</tr>
<tr>
<td></td>
<td>Flexor Carpi Radialis</td>
<td>Medial Epicondyle of Humerus</td>
<td>Base of 2nd &amp; 3rd Metacarpal Bone</td>
</tr>
<tr>
<td></td>
<td>Anconeus</td>
<td>Posterior Surface of Lateral Humerus</td>
<td>Lateral Margin of Olecranon &amp; Ulnar Shaft</td>
</tr>
</tbody>
</table>

The total elbow complex consists of several muscles which are responsible for elbow flexion-extension and forearm supination-pronation motion. It can be mentioned that the functions of these muscles are not unique. The same muscles are sometimes liable to generate different DOF motion. Usually, biceps and triceps muscles are considered as the prime mover for the elbow flexion and extension motion, respectively. It can be noted that the short head of biceps and the long head of biceps are also termed as proximal part of biceps and lateral part of biceps, respectively. Moreover, triceps long head and triceps lateral head are also termed as proximal part of triceps and lateral part of triceps, respectively. Brachialis and brachioradialis muscles are also responsible to flex the elbow and anconeus muscles are responsible to extend the elbow. However, as these are deep muscles, it is difficult to measure EMG signals.
from these muscles using surface electrodes. As a result, deep muscles are not considered in this thesis. Most of the muscles which move the elbow originate from the humerus and insert upon the forearm [34]. The forearm supination-pronation motion is also generated by the combined action of several muscles, such as the pronator teres, pronator quadratus, brachioradialis, anconeus, flexer carpi radialis, biceps, supinator, extensor carpi radialis longus and extensor carpi radialis brevis. Since all of these muscles are positioned only a limited space inside of the human arm, it is very difficult to distinguish the corresponding EMG signals using surface electrode on the outer skin surface. As a result, only eight kinds of muscles which are listed in Table 2.2 are considered for the control of the proposed prostheses. Three of them (pronator teres, proximal part of biceps and lateral part of biceps) are used for the elbow flexion motion and another two (proximal part of triceps and lateral part of triceps) are used for the elbow extension motion. Furthermore, three of them (pronator teres, flexer carpi radialis and anconeus) are used to determine the pronation motion and another three (extensor carpi radialis longus, proximal part of biceps and lateral part of biceps) are used to determine the supination motion. The locations of these muscles are shown in Fig. 2.10 to 2.12.

**FIGURE 2.10**: Elbow (a) Flexor and (b) Extensor Muscles.

**FIGURE 2.11**: Forearm Supinator Muscles.
FIGURE 2.12: Forearm Pronator Muscles.
Arthroplasty or joint replacement is a surgical procedure in which a damaged or arthritis affected articulation or joint is removed and replaced with an artificial joint or prosthesis. Prostheses are usually made for human main joints such as hip, knee, elbow, shoulder, wrist etc. Although in most of the cases the reason behind the arthroplasty is the arthritis pain, it’s not only the reason. The human articulations can be damaged by several reasons such as due to severe impacts or unusual stresses. However, the function of the conventional prostheses is to reduce the arthritis pain, improve the joint functions, and the stability [36] [37]. In this thesis, the concept of an implantable power assist robotic prosthesis (i.e., inner skeleton robot) is proposed that is supposed to substitute the existing conventional prostheses by incorporating an implantable actuator with the prosthesis to attain motion assist from inside of the body. At the first step toward the development of the implantable power assist robotic prosthesis, the elbow joint prosthesis is selected first due to its simpler mechanical design and anatomical shape than the complex knee and hip prostheses. The proposed prosthesis is expected to assist the joint motion from inside of the body in addition with meet the basic criteria fulfilled by the conventional prosthesis. It is obvious that the power assist prosthesis will become more attractive for the persons seeking for arthroplasty. This chapter summarizes the conventional elbow prothesis and the total elbow arthroplasty (TEA), and describes design and control strategy of the proposed implantable power assist prosthesis.
3.1 TOTAL ELBOW ARTHROPLASTY (TEA) AND CONVENTIONAL
ELBOW PROSTHESSES

The elbow joint replacement surgery or Total Elbow Arthroplasty (TEA) is nearing the end of its formative period as improvements in surgical technique and prosthetic design lead to consistent, reproducible results [36]. The primary indication for total elbow arthroplasty is pain relief. A secondary indication is to provide stability, with restoration of motion a third indication but rarely the primary one [37]. The artificial elbow joint or elbow prosthesis consists of two high quality metal stems which are generally termed as humeral and ulnar stem. These two stems are joined together with a metal/plastic hinge that allows the elbow to generate flexion-extension motion. The human bones have a comparatively soft, porous bone tissue in the centre which is commonly termed as “canal”. In elbow arthroplasty, special instruments are used to remove part of canal from the humerus and ulna bone, and two stems of the elbow prosthesis are inserted into each of the two prepared canals [38]. Bone cement is used to hold the stems firmly inside of the canals. Though the elbow replacement is a difficult and relatively uncommon procedure than that of the knee or hip prosthesis, with advances in prosthetic design, the incidence of complications is decreasing. There are 3 major types of elbow prosthesis: constrained, semiconstrained and unconstrained. Constrained prostheses have met with limited success and are no longer commonly used. Semiconstrained and unconstrained prosthesis are the most common devices being used today [87]. These prostheses are discussed briefly as bellows.

3.1.1 CONSTRAINED PROSTHESIS

TEA was first reported in 1972 by Dee using a constrained prosthesis in a patient with rheumatoid arthritis [37]. Constrained elbow prosthesis is just like a mechanical hinge joint which replaces the affected elbow joint. An example of constrained elbow prosthesis is shown in Fig. 3.1. The human elbow complex is not a completely mechanical hinge joint. During elbow flexion-extension motion it permits a slight axial rotation of the ulna with respect to the humerus and the amount of the rotation is about 5° [88]. Since the constrained elbow prosthesis does not allow any axial rotation, normal elbow motion results a large amount of stress on the prosthesis-bone
interface. The resultant forces act along the anterior and posterior directions with a magnitude of up to about 3 times the body weight and caused the bone cement to break down [89]. This leads the stem becoming loose, caused pain and finally requires revision. As a result, the constrained prosthesis has failed to gain wide acceptance.

![Constrained Elbow Prosthesis](image1.png)

**FIGURE 3.1:** Constrained Elbow Prosthesis [90].

### 3.1.2 SEMICONSTRAINED PROSTHESIS

The linked semiconstrained prostheses are much more versatile than unconstrained prosthesis [36]. The proposed implantable power assist prosthesis is designed to maintain the principle of semiconstrained prosthesis link design as those devices provide inherent stability to the joint because their linked designs significantly reduce the need for intact soft-tissue support. Semiconstrained prostheses allow some degree of freedom in axial rotation which in turns lowers the stresses developed on the bone-stem interface and consequently reduces the probability of loosening the stem. The GSB III Prosthesis is an example of the semiconstrained prosthesis as shown in Fig. 3.2.

![GSB III Semiconstrained Prosthesis](image2.png)

**FIGURE 3.2:** GSB III Semiconstrained Prosthesis.
3.1.3 **UNCONSTRAINED PROSTHESIS**

Unconstrained often called Nonconstrained or resurfacing prosthesis is designed for use in joints with intact soft-tissue support. It replicates the anatomy of the joint. An example of unconstrained prosthesis is shown in Fig. 3.3. This prosthesis consist a convex metal part and a concave polyethylene or metallic component part which are cemented inside of the humerus and ulna, respectively. This is not a hinge joint as there is no mechanical connection between the two parts. Since the humeral and ulnar components are not linked, this prosthesis provides an excellent joint stability. The implantable power assist prosthesis is proposed and designed for elderly or physically disabled persons. The intact soft tissue, ligament and musculature attachment can be rarely found to those people. However, for the patient seeking for arthroplasty with the soft tissue remain predominantly intact; this type of prosthesis can also be redesigned as implantable motion assist prosthesis.

![Unconstrained KUDO Prosthesis](image)

**FIGURE 3.3:** Unconstrained KUDO Prosthesis [36].

3.2 **IMPLANTABLE POWER ASSIST ROBOTIC PROSTHESIS (INNER SKELETON ROBOT) FOR HUMAN ELBOW MOTION**

The application of the elbow joint replacement surgery is commonly found among the aging society particularly among the persons suffered by arthritis pain at elbow complex. One of the major limits of these elderly persons in performing their daily life activity is their reduced muscles strength. The conventional elbow prostheses (artificial elbow joint) used in elbow joint replacement surgeries helps
those people by reducing their arthritis pain in addition with restoring their joint functions and stability. If the implantable elbow prosthesis could be able to assist the joint motion from inside of the body in addition with their basic functions, it is expected to become more attractive for the persons seeking for elbow arthroplasty.

In order to do so, an implantable 2 DOF power assist prosthesis that is anticipated to assist the elbow complex motions from inside of the body is proposed in this thesis. As the proposed prosthesis is a robotic device that is implanted inside of the skeleton of human body, it is termed as “inner skeleton robot”. In order to realize the natural motion assist from inside of the body, the approach is to design the inner skeleton robot that can imitate the functions provided by the normal elbow complex. Human elbow complex provides two DOF (elbow flexion-extension motion and forearm supination-pronation motion) with specific range of motion and the specific shape of the bony surfaces which form the articulations, keep the movement within the normal range. The implantable power assist system can be realized by replacing the bony surfaces of an articulation with the stator and rotor of an implantable joint actuator that is designed based on the shape and range of motion of that articulation.

The schematic diagram of the proposed 2 DOF inner skeleton robot to assist the elbow complex motions from inside of the human arm is shown in Fig. 3.4. This robot consists of an elbow actuator to assist elbow flexion-extension motion, a forearm actuator to assist forearm supination-pronation motion, and a T-mechanism. It is essential in prosthesis design to keep the limb motion similar to the normal human limb after the arthroplasty. In normal human elbow complex, the radius bone of the human forearm rotates and crosses over the ulna bone during pronation motion whereas during supination motion they lies just about parallel in position, which has discussed in chapter 2. A schematic diagram of the radius and ulna bone during supination-pronation motion has also shown in Fig. 2.6. In order to keep the similar bone arrangement in inner skeleton robot, the T-mechanism is proposed in this study. The T-mechanism, which consists of an elbow actuator rotor, two ball joints, radius and ulna stem, radius and ulna bone, T-link and the rotor of the forearm actuator, is designed to directly transmit the assist power from the forearm actuator to the radius and ulna bones and also to crosses the radius bone over the ulna bone like normal human elbow complex. In the proposed system, the stator of the elbow actuator is implanted on the distal part of the humerus bone and its rotor is attached with the proximal part of the forearm by means of radius and ulna stem as shown in Fig. 3.4.
The radius and ulna stems, which are implanted inside of the radius and ulna bones, respectively, are attached on the convex face of the elbow rotor by means of two ball joints. The human bones have a comparatively soft, porous bone tissue in the centre which is commonly termed as “canal”. In elbow arthroplasty, special instruments are used to remove part of canal from humerus and ulna bone, and two stems of the elbow prosthesis are inserted into each of the two prepared canals. Bone cement is used to hold the stems firmly inside of the canals. In the proposed system, the radius and ulna stems are expected to implant inside of the radius and ulna bones, respectively, like conventional elbow arthroplasty. In order to assist the supination-pronation motion, the forearm actuator is fixed on the convex side of the elbow rotor (between two stems) that generates motion in a perpendicular plane relative to the elbow actuator. The T-link has two holes at its two ends as shown in Fig. 3.4. The rotor (shaft) of the forearm actuator is fixed on the base of the T-link, and the radius and ulna bones pass through the holes of the T-link. Therefore, the T-link is able to rotate freely on the surface of each bone. Since the base of the T-link is fixed with the shaft of the forearm actuator and its ends are attached to the two bones, rotation of the
forearm actuator makes one end of the T-link move upward and the other end move downward as shown in Fig. 3.4, so that the movement of pronation or supination is generated. The diameters of the T-link end holes are made to be a little larger than the bones sizes, as the bones move inside of the holes during the forearm motion. Moreover, both of the bone surfaces and the holes are supposed to be covered with a protective layer (i.e., artificial cartilage) to reduce the risk of wear and pain. The direction of rotation of the T-link ends during forearm supination-pronation motion is shown in Fig. 2, by dotted and solid line, respectively. Practically, the forearm actuator and the T-link should be small enough to safely place between the muscles of the bones. The thickness and the width of the T-link between the end holes should be designed to be minimal sizes and a highly finished surface with smooth edge is expected to avoid damaging of surrounding muscles as it rotates inside of the arm during the forearm motion. Moreover, it is expected to connect the T-link with the radius and ulna bone in such a position where the movement of the T-link does not interact with the surrounding biological tissues such as muscles.

3.3 Prototype 2 DOF Inner Skeleton Robot

**Figure 3.5:** Prototype 2 DOF Inner Skeleton Robot.
The prototype 2 DOF inner skeleton robot developed in this study to realize the proposed motion assist system is shown in Fig. 3.5. Usually, the limitation of the movable range of the elbow flexion-extension motion is 145 degrees in flexion and 5 degrees in extension and that of the forearm supination-pronation motion is 80-90 degrees in supination and 50-80 degrees in pronation. Considering the safety in the proposed system, the limit of the elbow motion is designed to provide 115 degrees in flexion and 0 degree in extension and that of the forearm motion is 55 degrees in both supination and pronation motion. A small size cylinder (26 mm diameter with 50 mm height) made up with nonmagnetic material is used in the prototype robot as a jig to hold the stator of the elbow actuator firmly. A shaft is placed in the centre of the cylinder by means of two ball bearings. The elbow actuator rotor is held up in front of the stator (the convex stator pole faces the concave side of the rotor, i.e., movable plate) with an aluminum bar and the end of the bar is fixed with the center shaft. This makes the elbow actuator to provide hinge motion as like the elbow joint [91]. In this study, an example of implantable joint actuator is developed for the elbow joint to assist the elbow flexion-extension motion and a DC servo motor (RH-5A-5502-E03640, Harmonic Drive) has been used for forearm supination-pronation motion. However, the proposed implantable actuator can be designed for both of the elbow and forearm motion. Although the ends of the T-link are attached with the radius and ulna bone in the proposed system, comparatively lengthy stems are used and the ends of the T-link are connected with the stems to verify the effectiveness of the T-mechanism in the prototype robot. Moreover, the stems are placed in a horizontal plane for the ease of demonstration, although they are supposed to be placed in a vertical plane.

3.4 IMPLANTABLE JOINT ACTUATOR

3.4.1 DESIGN OF THE ACTUATOR

At the first step toward the development of the proposed implantable power assist system, it is necessary to develop an actuator that is able to act like a human articulation and assist the motion from inside of the body. The implantable joint actuator has to be designed according to the joint shape and the joint range of motion. In this study, an example of implantable joint actuator for the elbow joint to assist the
elbow flexion-extension motion is developed. The schematic diagram of the actuator is shown in Fig. 3.6.

**Figure 3.6:** Schematic Diagram of the Joint Actuator.

Unlike the conventional actuator, the designed actuator does not provide complete rotary motion as the movable range available at the elbow joint is limited by the bony geometry as well as by the resistance provided by the soft tissues. Thus the range of motion of the actuator (i.e., the movable range of the proposed system for the elbow motion) is limited according to the elbow joint motion. Considering the factor of safety, the actuator is designed to provide 115° in flexion and 0° in extension motion. Several kinds of actuation systems such as electromagnetic actuator, switched reluctance actuator, or ultrasound actuator can be considered as a good candidate for the actuator of the inner skeleton robot [92]. Switched reluctance actuators are simpler, more robust in construction than the traditional electromagnetic actuator [93]. The electromagnetic actuator consists of complex mechanical structure, more expensive, and required maintenance than the switched reluctance actuator [43]. As a result, a simple, low-cost direct drive motion switched reluctance actuator has been designed as an example of joint actuator for the proposed implantable motion assist system.

The actuator consists of two parts; a stator and a rotor. The stator (fixed part) and the rotor (movable part) is designed so that they can be safely attached with distal part of humerus bone and proximal part of forearm respectively, like elbow arthroplasty and generate human like motion. Since the design of the actuator is different from the traditional rotary actuator, the absence of permanent magnet in the actuator reduces the possibility to interact with the surrounding environment, so that it is suitable to
implant inside of the body. The stator and rotor of the actuator have machined from 0.5 mm sandwiched silicon laminated steel “50A700” plate. The stator part of the actuator consists of 20 poles in two rows; 10 poles in upper row and 10 poles in the lower row and the combination of the upper and lower row poles makes 10 exciting phase. The coils are wound around the poles in such a way that at each exciting sequence 1 upper row pole and its consecutive 1 lower pole will generate two opposite magnetic pole to form one exciting phase. The radius of the convex shape stator pole face is 25 mm and the radius of the concave face of the rotor (the movable plate) is kept 26 mm and thus maintains 1 mm intermediate space between the fixed and the movable part of the actuator. The length of each upper and lower stator pole face is 10 mm and the distance between the upper and lower pole face is also kept at 10 mm. The rotor is just a 75 degree curve plate with 30 mm length and 5 mm thickness. The design of the stator and rotor are shown in Fig. 3.7 and 3.8, respectively.

**Figure 3.7:** Design of the Stator.

**Figure 3.8:** Design of the Rotor.
3.4.2 Advantages & Disadvantages of the Designed Actuator

The designed actuator as well as the switched reluctance actuators offers numerous advantages over than the electromagnetic actuator. These can be outlined as:

- **Performance** – Offers greater amount of torque.
- **Compactness** – Small compact unit size makes efficient use of material. Moreover, the direction of torque does not depend on the direction of current i.e. the direction of current is immaterial. Thus small semiconductor can be used in lieu of relatively large extra hardware (like motor driver) that is necessary to change the direction of current for the electromagnetic actuator.
- **Magnet** – Absence of permanent magnet makes it suitable for implantation inside of the human body.
- **Cost** – Switched reluctance actuator is cheaper than electromagnetic actuator.
- **Heating Problem** – The amount of heat generation is less than that of electromagnetic actuator.

The disadvantages of the actuator can be outlined as follows:

- **Control** – The control of the actuator is not simple.
- **Noise** – The magnetic flux across the air gap between the stator and rotor creates noise.

3.4.3 Operating Principle of the Designed Actuator

The designed actuator must be operated in a continuous switching mode to generate the desired motion. Sequentially switching the stator coils caused the rotor to rotate. The movement of the rotor, hence the production of torques, involves switching of current into stator winding when there is a variable of reluctance. As a result the actuator is named as “Switched Reluctance Actuator”. The design of the actuator assures a simple stator excitation sequence and the direction of motion (as well as the direction of the rotor) can be reverse by simply changing the sequence of the stator excitation. The basic working principle to rotate the rotor is to excite the stator phase that exists nearest aligned position of the rotor. The excitation sequence
of the stator phases for clockwise (CW) and counterclockwise (CCW) rotation of the rotor is shown in Fig. 3.9 and 3.10, respectively. Since the forearm is supposed to attach with the rotor of the actuator, fully extended and flexed position of the forearm can be realized with the stator-rotor combination as shown in Fig. 3.9(a) and 3.10(a), respectively. At the extended position exciting sequence 5-6-7-8-9-10 will result the CW (flexion) motion of the movable plate whereas exciting sequence 6-5-4-3-2-1 will result the CCW (extension) motion of the movable plate from its flexed position. At the extended position, consecutive excitation of the stator phases 5, 6, 7, 8, 9, and 10, will generate the CW rotation of the rotor as shown in Fig. 3.9(a)-(g). On the other hand, consecutive excitation of the stator phases 6, 5, 4, 3, 2, and 1, will rotate the rotor towards the CCW direction from the flexed position as shown in Fig. 3.10(a)-6(g). When the stator poles are at unaligned position, the poles are excited by supplying current and the current supply is terminated after reaching the aligned position. The stator poles and the movable plate have been designed to provide about 19 degree step increments when current applied to the appropriate sequence on the stator phases. Since the direction of current is immaterial for the direction of rotation, relatively inexpensive and small size semiconductors are used to control the actuator.

Considering a linear magnetic characteristics of the actuator, the torque produced by the movable plate can be calculated by the Eq. (3.1).

\[ T_e = \frac{1}{2} i^2 \frac{dL}{d\theta} \]  

(3.1)

where \( L \), \( \theta \) and \( i \) are inductance, rotor position, and current, respectively. Equation (3.1) can be written as:

\[ T_e = \frac{1}{8} k_1 \frac{N_p^2}{l_g} i^2 \]  

(3.2)

where \( k_1 \) is a constant and, \( N_p \) and \( l_g \) are the number of turns of stator coil and air gap length between stator and rotor, respectively. It can be observed from Eq. (3.2) that the torque is proportional to the square of current and square of number of turns of stator coil, and inversely proportional to the air gap length.
FIGURE 3.9: Stator Phase Exciting Sequence to Generate Clockwise Motion.

FIGURE 3.10: Stator Phase Exciting Sequence to Generate Counterclockwise Motion.
3.4.4 ANGULAR POSITION SENSOR FOR THE DESIGNED ACTUATOR

Torque of the designed actuator is controlled based on the human muscles EMG signals. As a result, the controlled amount of input current is applied on the respective stator phases according to the rotor position to obtain clockwise and counterclockwise direction that demand the necessity of an angular position sensor to control the actuator. The conductive plastic element (CPE) is used here to design the angular position sensor to control the actuator. The CPE is used as the resistive material and attached with the upper row stator poles and a wiper is attached with the movable plate as shown in Fig. 3.11. The shape of the conductive plastic element is designed according to the shape of the stator part of the actuator. Applying a small amount of constant voltage to the ends of the CPE generates variable voltage at its conductive layer face and the amount of generated voltage is proportional to the angular distance of the conductive layer face. The wiper is attached with the movable plate in a way that during the movement of the movable plate the wiper travels across the conductive layer face to pick up the voltage. The position of the movable plate is determined by measuring the voltage value picked up by the wiper.

![Joint Actuator with Angular Position Sensor](image)

**FIGURE 3.11**: Joint Actuator with Angular Position Sensor.
3.5 **HUMAN MUSCLES ELECTROMYOGRAM (EMG) SIGNALS**

The human arm muscles electromyogram (EMG) signals are used as controller input information for the control of the prostheses proposed in this thesis. Electromyogram is one of the most important biological signal in which the human motion intention is directly reflected. Therefore, EMG signals are often used as controller input information for a robot system [94]-[96]. The EMG signals represent the amount of electrical charge generated in the muscles fibers.

### 3.5.1 Characteristics of EMG Signals

The amplitude of the EMG signals is usually stochastic (random) in nature and can be reasonably represented by Gaussian distribution function. The amplitude of the signal can varied from 0 to 10 mV (peak to peak). EMG signals are used as potential input information to control the electromechanical active joint of the robotic system. However, complexity arises due to inherent instability nature of the EMG signals. It is very difficult to obtain the same EMG signals for the same motion even from the same person. Some typical events that are responsible for the random nature of EMG signals are listed below [97].

- Activity level of each muscle and the way of using each muscle for certain motion is different between persons.
- Since responsibility of each muscle for the motion varies in accordance with joint angles, activity level of each muscle is highly non-linear in nature.
- It is usual that one muscle is not only concerned with one motion rather multiple motions, and therefore shows the open loop characteristics.
- Activity level of bi articular muscles is affected by the motion of other joint, since load acting on other joint affects the activity level of them.
- Human physiological conditions such as fatigue, stress and others affects the activity level of muscles.
- In addition to all these problems, inherent noise in the electronics components in the detection and recording equipment, ambient noise, motion artifacts and others affects the activity level of muscles.
3.5.2 **Skin Surface EMG Signals Detection Procedure**

The effectiveness of the prostheses proposed in this thesis has been evaluated by using the skin surface EMG signals. To be used skin surface EMG signals as control command signals, it is highly desirable to measure the EMG signals that contain the maximum amount of information of the subject motion intention and the amount of contaminated electrical noise should be as low as possible. Thus, the maximization of the signal-to-noise ratio should be done with minimal distortion to the EMG signals [98]. In this study, “10mm Ag/AgCl NE-121J, Nihon Kohden” surface electrodes were used to record the EMG signals from subjects arm muscles. The step-by-step procedures adopted for the recording of EMG signals can be outlined as follows:

- At the first step, subject skin and surface electrodes should be cleansed with an antiseptic solution (usually alcohol) to remove dirt.
- Dry skin provides insulation from static electricity and others. To avoid this, a conductive ionic gel is used between the surface electrode and the skin surface instead of directly placing the electrode on the dry skin.
- Unwanted movement of the electrode on the skin surface generates a great amount of noise. In order to reduce the amount of noise, adhesive tapes are used to place the surface electrodes firmly on the skin surface. It can be noted that, self adhesive surface electrode can also be used in which there is no necessity of using extra adhesive tape.
- The surface electrode pairs should be placed in parallel with muscles fibers. It is usually recommended to place the electrode pair about 10mm apart [98].
- The ground electrode should be placed as far away as possible from the muscles being measured and should be place on electrically neutral tissue.

3.5.3 **Feature of EMG Devices**

The amount of electrical discharge by the human body muscles as well as the magnitude of the EMG signals is a very minute amount to be used directly as the controller input information. In order to be used the surface EMG signals to evaluate the effectives of the proposed prostheses, biofeedback devices - Input Box (JB-620J, Nihon Kohden), Multi Channel Amplifier (MEG 6108, Nihon Kohden) and Evoked
Surface Electrodes (10mm Ag/AgCl NE-121J, Nihon Kohden) are used in combination to convert the EMG signals to a meaningful amount. The flowchart shown in Fig. 3.12 demonstrates the procedure adopted in this study for recording the surface EMG signals. The amount of the EMG signals picked up by the surface electrodes are firstly send to the input box from which finally send to the amplifier for necessary amplification. The signal amplifier offers an adjustable gain setting. Appropriate adjustment of the gain settings is an important feature of effective EMG recording. On low gain settings the machine will require a greater signal before the output changes. On higher gain settings, a small amount of EMG activity will be easily seen by the user. In this study, the “Gain” for the multi channel amplifier is set to $50 \mu V/V$. The amplified EMG signals are then send to the AD converter and sampled at a rate of 2 KHz.

**Figure 3.12**: Skin Surface EMG Signals Recording Technique.
3.5.4 **PROCESSING RAW EMG SIGNALS**

EMG signals consist of a wide frequency range and it is sometimes very difficult to reduce noise by filtering techniques. Furthermore, it is difficult to use raw EMG signals as the input information for the controller of the robotic devices. Therefore, it is essential to extract the feature from the raw EMG signals. There are many kinds of feature extraction methods, for example, mean absolute value, mean absolute value slope, zero crossings, waveform length or root mean square [99]. In this study, root mean square (RMS) values are preferred to process raw EMG signals (0.01-10mV, 0.01-2 KHZ). The RMS value is a measure of power of the signal and is widely used in most applications as the calculation can be expected to be carried out almost in real time [12]. The equation of RMS value is written as:

\[
RMS = \sqrt{\frac{1}{N} \sum_{i=1}^{N} v_i^2}
\]

where \(v_i\) is the voltage value at \(i^{th}\) sampling, and \(N\) is the number of samples in a segment. The number of samples is set to be 100 and the sampling time is set to be 500 microseconds in this study.

![Raw EMG RMS EMG](image)

**FIGURE 3.13**: Conversion of Raw EMG Signals into RMS Values.

3.6 **CONTROL STRATEGY FOR 2 DOF INNER SKELETON ROBOT**

A fuzzy-neuro control method has been used to control the proposed prototype of implantable power assist prosthesis or inner skeleton robot which enables the cooperative motion of the human elbow and forearm motion [15]. The inner skeleton
robot is supposed to be controlled based on the activation pattern of the intramuscular electromyogram signals of patient’s muscles that directly reflect the patient motion intention in order to realize natural motion assist. Intramuscular EMG signals are commonly measured using either fine wire electrodes inserted into the muscle by means of hypodermic needles or by needle electrodes [33]. Implantable Titanium coated flexible wire with needle tip can be expected to measure the intramuscular EMG signals for the proposed implantable motion assist system, though much research is required to find out the suitable way to measure the intramuscular EMG signals for the proposed motion assist system. The human forearm and upper arm consists of a number of muscles in a limited space inside of the arm which are involved in many motions. Consequently, the same muscle is sometimes used for different motions. Although a number of muscles are responsible for the elbow and forearm motion, eight kinds of EMG signals (ch.1: pronator teres, ch.2: flexer carpi radialis, ch.3: anconeus, ch.4: supinator, ch.5: medial part of biceps, ch.6: lateral part of biceps, ch.7: medial part of triceps and ch.8: lateral part of triceps) are measured for control of the forearm supination-pronation motion and elbow flexion-extension motion in this study. Three of them (ch.1: pronator teres, ch.5: medial part of biceps and ch.6: lateral part of biceps) are used for the elbow flexion motion and another two (ch.7: medial part of triceps and ch.8: lateral part of triceps) are used for the elbow extension motion. Furthermore, three of them (ch.1: pronator teres, ch.2: flexer carpi radialis and ch.3: anconeus) are used to determine the pronation motion and another three (ch.4: supinator, ch.5: medial part of biceps and ch.6: lateral part of biceps) are used to determine the supination motion. Thus, the biceps is used for both the forearm supination motion and elbow flexion motion and pronator teres is used for forearm pronation motion and elbow flexion motion. It can be noted that the anconeus muscles is also acts as an elbow extensor and both of the ch.3: anconeus and ch.4: supinator is also affected by the grasping force of the hand. The position of the surface electrodes to measure the EMG signals from the designated muscles is shown in Fig. 3.14. The amplified EMG signals are sampled at a rate of 2 KHz and its root mean square value is calculated as shown in Eq. 3.3.

The initial fuzzy IF-THEN control rules are designed based on the analyzed human forearm and elbow motion patterns in a pre-experiment and then transferred to the neural network form to be a fuzzy-neuro controller. The architecture of the fuzzy-neuro controller is depicted in Fig. 3.15.
**Figure 3.14**: Position of Surface Electrodes.

**Figure 3.15**: Architecture of the Fuzzy-Neuro Controller.
The input variables of the fuzzy-neuro controller are the eight RMS values of the EMG signals and the outputs are the torque command for the forearm supination-pronation motion \( \tau \), the desired impedance parameters \( (B_e, K_e) \) is the damping coefficient and \( K_e \) is the spring constant) and the desired angle \( (q_d) \) for the elbow flexion-extension motion of the proposed actuator. In the architecture of the fuzzy-neuro controller, \( \sum \) means the summation of the inputs and \( \prod \) means the multiplication of the inputs. Three kinds of fuzzy linguistic variables (ZO: zero, PS: positive small and PB: positive big) are prepared for the RMS of 8 EMG signals as depicted in Fig. 3.16 and 3.17. Two kinds of nonlinear functions \( (f_G, f_S) \) are used to express the membership function of the neuro-fuzzy controller:

\[
f_G(u_G) = \frac{1}{1 + e^{-\omega o}} \\
u_s(x) = \omega_o + \omega_i x \\
f_G(u_G) = e^{-\omega o} \\
u_G(x) = \frac{\omega_o + x}{\omega_i}
\]

where, \( x \) is the input signal, \( \omega_o \) is a threshold value and \( \omega_i \) is a weight.

The weight value, \( w_i \) and threshold value, \( w_o \) for the ZO membership function are calculated as follows:

\[
w_i = (\log((1.0 / H_RATE - 1)/(1.0 / L_RATE - 1)))/(w_3 - w_1) \\
w_o = ((-1.0) \log(1.0 / H_RATE - 1.0) - w_i w_i
\]

For the PS membership function, \( w_i \) and \( w_o \) are calculated as:

\[
w_i = \sqrt{(-w_3 + (w_3 + w_3)/2)} / \log(1.0 / M_RATE) \\
w_o = -w_3
\]

For the PB membership function, \( w_i \) and \( w_o \) are calculated as:

\[
w_i = (\log((1.0 / H_RATE - 1)/(1.0 / L_RATE - 1)))/(w_3 - w_3) \\
w_o = ((-1.0) \log(1.0 / H_RATE - 1.0) - w_i w_3
\]

The process of the fuzzy-neuro controller is the same as that of ordinal simplified fuzzy controllers. Consequently, the output of the fuzzy-neuro controller is calculated with the following equation:
FIGURE 3.16: Membership Function of RMS Value (EMG) for Elbow Flexion-Extension Motion.
**FIGURE 3.17**: Membership Function of RMS Value (EMG) for Forearm Supination-Pronation Motion.
\[
O = \frac{\sum \omega_i y_{ki}}{\sum y_{ki}} \quad (3.14)
\]

where,
- \( O \) = Output vector;
- \( y_{ki} \) = Degree of fitness of \( i_{th} \) rule;
- \( \omega_i \) = Weight for \( i_{th} \) rule;

In the fuzzy-neuro controller, 15 kinds of fuzzy IF-THEN rules are used to generate the desired elbow joint angle and impedance for the actuator designed for elbow flexion-extension motion and 17 kinds of fuzzy IF-THEN rules are prepared to generate the desired torque for the forearm supination-pronation motion actuator. The IF-THEN rules for elbow and forearm motions are listed in Table 3.1 and 3.2, respectively. The fuzzifier layer, in which each neuron represents a membership function for the input, consists of 24 neurons, the rule layer consists of 32 neurons and the defuzzifier layer consists of 5 neurons. The controller output torque command for the forearm actuator is then transferred to the DC motor through the motor driver (PC-0121-2 Titech Driver), which is then finally transferred to the radial and ulnar stem through the T-mechanism. Impedance control is performed with the derived impedance parameters and the derived desired angle for the elbow joint control of the inner skeleton robot [12], [100]-[101]. The torque of the proposed actuator \( (\tau_e) \) for elbow flexion-extension motion is calculated using the following equation:

\[
\tau_e = M_e \ddot{q}_d + B_e (\dot{q}_d - \dot{q}) + K_e (q_d - q) + \tau_s \quad (3.15)
\]

where, \( M_e \) is the moment of inertia of the forearm and \( \tau_s \) is the external torque which is zero if there is no external load on the forearm and \( q \) is the angle of the actuator measured by the designed position sensor. The elbow torque command is then converted into the voltage value and applied to the base of the semiconductor that are used to control the current in the stator phase. In this study, 10 TIP-102 NPN transistors are used to control the current in 10 stator phases. The torque command as well as the voltage command is applied to the base of the transistor as the base current controls the current flow between the collector and the emitter of the transistor. It can be noted that, the excitation of the appropriate stator phase (i.e., selection of the transistor) will be depend upon the position of the rotor (movable plate).

The EMG based control rules are sometimes different when the arm posture is changed since role of each muscle is changed according to the arm posture. To
**TABLE 3.1:** Fuzzy IF-THEN Control Rules for Elbow Flexion-Extension Motion.

<table>
<thead>
<tr>
<th>Rule</th>
<th>IF</th>
<th>THEN</th>
</tr>
</thead>
<tbody>
<tr>
<td>01</td>
<td>EMG Ch.1 is PB, EMG Ch.2 is PS, and EMG Ch.3 is PS</td>
<td>$q_d = 0.001000$, $B_e = 0.453$, $K_e = 1.0$</td>
</tr>
<tr>
<td>02</td>
<td>EMG Ch.1 is PB, EMG Ch.2 is PB, and EMG Ch.3 is PB</td>
<td>$q_d = 0.002618$, $B_e = 1.359$, $K_e = 9.0$</td>
</tr>
<tr>
<td>03</td>
<td>EMG Ch.1 is PB, EMG Ch.2 is PS, and EMG Ch.3 is PB</td>
<td>$q_d = 0.001570$, $B_e = 0.906$, $K_e = 4.0$</td>
</tr>
<tr>
<td>04</td>
<td>EMG Ch.1 is PB, EMG Ch.2 is PB, and EMG Ch.3 is PS</td>
<td>$q_d = 0.001570$, $B_e = 0.906$, $K_e = 4.0$</td>
</tr>
<tr>
<td>05</td>
<td>EMG Ch.4 is PS and EMG Ch.5 is PS</td>
<td>$q_d = -0.001000$, $B_e = 0.453$, $K_e = 1.0$</td>
</tr>
<tr>
<td>06</td>
<td>EMG Ch.4 is PB and EMG Ch.5 is PB</td>
<td>$q_d = -0.002618$, $B_e = 1.359$, $K_e = 9.0$</td>
</tr>
<tr>
<td>07</td>
<td>EMG Ch.4 is PS and EMG Ch.5 is PB</td>
<td>$q_d = -0.001570$, $B_e = 0.906$, $K_e = 4.0$</td>
</tr>
<tr>
<td>08</td>
<td>EMG Ch.4 is PB and EMG Ch.5 is PS</td>
<td>$q_d = -0.001570$, $B_e = 0.906$, $K_e = 4.0$</td>
</tr>
<tr>
<td>09</td>
<td>EMG Ch.2 is PS, EMG Ch.3 is ZO, and EMG Ch.4 is ZO</td>
<td>$q_d = 0.000000$, $B_e = 0.906$, $K_e = 4.0$</td>
</tr>
<tr>
<td>10</td>
<td>EMG Ch.2 is PB, EMG Ch.3 is ZO, and EMG Ch.4 is ZO</td>
<td>$q_d = 0.000000$, $B_e = 1.359$, $K_e = 9.0$</td>
</tr>
<tr>
<td>11</td>
<td>EMG Ch.2 is ZO, EMG Ch.3 is ZO, EMG Ch.4 is ZO, and EMG Ch.5 is ZO</td>
<td>$q_d = 0.001000$, $B_e = 0.000$, $K_e = 0.0$</td>
</tr>
<tr>
<td>12</td>
<td>EMG Ch.1 is PS, EMG Ch.2 is PS, and EMG Ch.3 is PS</td>
<td>$q_d = 0.001000$, $B_e = 0.453$, $K_e = 1.0$</td>
</tr>
<tr>
<td>13</td>
<td>EMG Ch.1 is PS, EMG Ch.2 is PB, and EMG Ch.3 is PB</td>
<td>$q_d = 0.002618$, $B_e = 1.359$, $K_e = 9.0$</td>
</tr>
<tr>
<td>14</td>
<td>EMG Ch.1 is PS, EMG Ch.2 is PS, and EMG Ch.3 is PB</td>
<td>$q_d = 0.001570$, $B_e = 0.906$, $K_e = 4.0$</td>
</tr>
<tr>
<td>15</td>
<td>EMG Ch.1 is PS, EMG Ch.2 is PB, and EMG Ch.3 is PS</td>
<td>$q_d = 0.001570$, $B_e = 0.906$, $K_e = 4.0$</td>
</tr>
</tbody>
</table>
Table 3.2: Fuzzy If-Then Control Rules for Forearm Supination-Pronation Motion.

<table>
<thead>
<tr>
<th>Rule</th>
<th>IF</th>
<th>THEN</th>
</tr>
</thead>
<tbody>
<tr>
<td>01</td>
<td>EMG Ch.1 is PB, EMG Ch.2 is PB, EMG Ch.3 is PB, EMG Ch.5 is ZO and EMG Ch.6 is ZO</td>
<td>Actuator Torque = 4.0 Nm</td>
</tr>
<tr>
<td>02</td>
<td>EMG Ch.1 is PB, EMG Ch.3 is PS, EMG Ch.5 is ZO and EMG Ch.6 is ZO</td>
<td>Actuator Torque = 2.0 Nm</td>
</tr>
<tr>
<td>03</td>
<td>EMG Ch.1 is PS, EMG Ch.3 is PS, EMG Ch.4 is PB, EMG Ch.5 is ZO and EMG Ch.6 is ZO</td>
<td>Actuator Torque = 2.0 Nm</td>
</tr>
<tr>
<td>04</td>
<td>EMG Ch.1 is PS, EMG Ch.3 is PS, EMG Ch.5 is ZO and EMG Ch.6 is ZO</td>
<td>Actuator Torque = 2.0 Nm</td>
</tr>
<tr>
<td>05</td>
<td>EMG Ch.1 is PS, EMG Ch.3 is PB, EMG Ch.5 is ZO and EMG Ch.6 is ZO</td>
<td>Actuator Torque = 2.0 Nm</td>
</tr>
<tr>
<td>06</td>
<td>EMG Ch.1 is ZO, EMG Ch.2 is ZO, EMG Ch.4 is PB, EMG Ch.5 is PB and EMG Ch.6 is PB</td>
<td>Actuator Torque = -4.0 Nm</td>
</tr>
<tr>
<td>07</td>
<td>EMG Ch.1 is ZO, EMG Ch.2 is ZO and EMG Ch.5 is PS</td>
<td>Actuator Torque = -2.0 Nm</td>
</tr>
<tr>
<td>08</td>
<td>EMG Ch.1 is ZO, EMG Ch.2 is ZO, EMG Ch.4 is ZO, EMG Ch.5 is PB and EMG Ch.6 is PB</td>
<td>Actuator Torque = -2.0 Nm</td>
</tr>
<tr>
<td>09</td>
<td>EMG Ch.1 is ZO, EMG Ch.2 is ZO, EMG Ch.5 is PB and EMG Ch.6 is PS</td>
<td>Actuator Torque = -2.0 Nm</td>
</tr>
<tr>
<td>10</td>
<td>EMG Ch.1 is ZO, EMG Ch.2 is ZO, EMG Ch.5 is PB and EMG Ch.6 is PB</td>
<td>Actuator Torque = -2.0 Nm</td>
</tr>
<tr>
<td>11</td>
<td>EMG Ch.1 is ZO, EMG Ch.2 is ZO, EMG Ch.5 is ZO and EMG Ch.6 is ZO</td>
<td>Actuator Torque = 0.0 Nm</td>
</tr>
<tr>
<td>12</td>
<td>EMG Ch.1 is PS, EMG Ch.3 is ZO and EMG Ch.6 is ZO</td>
<td>Actuator Torque = 2.0 Nm</td>
</tr>
<tr>
<td>13</td>
<td>ZO EMG Ch.1 is PB, EMG Ch.3 is PB, EMG Ch.5 is ZO and EMG Ch.6 is ZO</td>
<td>Actuator Torque = 3.0 Nm</td>
</tr>
<tr>
<td>14</td>
<td>EMG Ch.1 is PS, EMG Ch.3 is PS, EMG Ch.4 is PS, EMG Ch.5 is ZO and EMG Ch.6 is ZO</td>
<td>Actuator Torque = 2.0 Nm</td>
</tr>
<tr>
<td>15</td>
<td>EMG Ch.1 is ZO, EMG Ch.2 is ZO, EMG Ch.4 is PS, EMG Ch.5 is PB and EMG Ch.6 is PB</td>
<td>Actuator Torque = -3.0 Nm</td>
</tr>
<tr>
<td>16</td>
<td>EMG Ch.1 is PB, and EMG Ch.5 is PB</td>
<td>Actuator Torque = 2.0 Nm</td>
</tr>
<tr>
<td>17</td>
<td>EMG Ch.1 is PB, and EMG Ch.5 is PS</td>
<td>Actuator Torque = 2.0 Nm</td>
</tr>
</tbody>
</table>
overcome from these problems, multiple fuzzy-neuro controllers have been designed and applied under certain arm posture region. Based on the joint angles, the movable range of elbow flexion-extension motion is divided into three regions (FA: flexed angle, IA: intermediate angle and EA: extended angle). A membership function is defined for each region to switch the fuzzy-neuro controller smoothly. The membership functions for the joint angle are depicted in Fig. 3.18. This membership function acts as a biological switch for moderately selecting the appropriate controllers in accordance with the arm posture. It is to be noted here, a maximum of two kinds of neuro-fuzzy controller might be used during elbow flexion-extension motion. The structure of the proposed controller for the implantable 2 DOF inner skeleton robot is shown in Fig. 3.19.

**FIGURE 3.18**: Membership Function for Joint Angle.

**FIGURE 3.19**: Structure of the Proposed Controller.
3.7 Experiments

Experiments have been carried out to evaluate the measurement accuracy of the designed angular position sensor and the effectiveness of the inner skeleton robot by using EMG signals. A schematic diagram of the experimental setup is shown in Fig. 3.20. In the experimental setup, for ease of experiment, the CPE is fitted on the top of the cylinder that holds the stator and the wiper is connected with the Al bar that holds the rotor in such a way that during the movement of the rotor the wiper travels across the conductive layer face of the CPE. A second order low pass filter (cut off frequency 8 Hz) is used to reduce the noise from the voltage signal picked up by the wiper. The sampling time was set to be 500 microsecond for every experiment. The measurement accuracy of the designed angular position sensor is compared with an incremental rotary encoder (NOM-S600-2MC, Nemicon). The encoder is attached with the center shaft of the cylinder. As the rotor is attached with the center shaft by means of the Al bar, the amount of shaft rotation indicates the rotor position. In order to verify the measurement accuracy of the designed sensor for any velocities of the actuator in this experiment, the evaluation is performed under three different angular...
FIGURE 3.21: Comparison of Measurement Accuracy of the Designed Sensor with a Rotary Encoder.
velocities. At the same time, the angular position of the rotor is measured by the encoder for comparison. The experimental results for a low, medium and high velocity of the actuator are shown in Fig. 3.21(a), (b) and (c), respectively. The experimental results indicate that the proposed angular position sensor can measure the rotor position properly for each case.

In order to verify the performance of the proposed system by using EMG signal, experiment has been carried out with a healthy human male subject (29 years old). Since the intramuscular EMG signals require skin puncture, for ease of experiment in this study, experiment was carried out by using surface EMG signals. The effectiveness of the designed elbow joint actuator has examined based on the human elbow flexion-extension motion. Although the ultimate goal is to implant the actuator inside of the arm, the effectiveness of the actuator is verified outside of the human body. In the experiment, similar condition is maintained for both of the subject’s arm and for the actuator. Here, the subject carried out elbow flexion-extension motion without carrying any external load in his hand and the actuator is activated without any external load applied on the rotor also. If the subject carries an external load in his hand, the activity level of the muscles increases in accordance with the amount of the load and that results in increase of generating torque of the actuator. Therefore, the similar results would be obtained if the subject carries an external load in his hand. As the EMG signals vary for different arm motions, the experiment is performed for slow and fast elbow motions in order to verify the controllability of the actuator for different levels of EMG signals. The experimental results for slow and fast arm motions are shown in Fig. 3.22 and 3.23, respectively. Since the medial part of biceps and lateral part of triceps are the most active muscles for elbow flexion and extension motion respectively, only the EMG signals of channel 5 and channel 8 are depicted in Fig. 3.22 and 3.23. The experimental results show that the flexion motion is generated when the biceps muscles are activated and consequently extension motion is generated when the triceps muscles are activated. Moreover, the actuator can follow the desired joint angle (trajectory) generated by the fuzzy-neuro controller (i.e., one of the output of the fuzzy -neuro controller: $q_d$).

The effectiveness of the proposed prototype of 2 DOF inner skeleton robot for a combined motion of the elbow flexion-extension and forearm supination-pronation motions has also been evaluated. In this experiment, the subject is performed the cooperative motion of the elbow and forearm (i.e., a combination of elbow flexion-
**Figure 3.22**: Experimental Results of the Actuator – Slow Arm Motion.

**Figure 3.23**: Experimental Results of the Actuator – Fast Arm Motion.
extension motion and forearm supination-pronation motion) without any external load in his hand and the inner skeleton robot is allowed to generate motions without any external load applied on the rotor of the actuators also. The similar results would be obtained if the subject carries an external load in his hand, since the activity level of the muscles is increased in accordance with the amount of the external load. The experimental results for slow and fast arm motion are shown in Fig. 3.24 and 3.25, respectively. The biceps muscles are activated during both the elbow flexion motion and forearm supination motion. The pronator teres muscle is responsible for the forearm pronation motion as well as for the elbow flexion motion. And the activation of the triceps muscles results the elbow extension motion. Therefore, the same muscle is sometimes used for both of the flexion-extension and supination-pronation motion. The fuzzy-neuro controller is designed to recognize the effect of the muscles which are common to both of the flexion-extension and supination-pronation motion. Although eight EMG signals are used to control the inner skeleton robot, for the ease of demonstration only three EMG signals (lateral part of biceps, lateral part of triceps and pronator teres) are depicted in Fig. 3.24 and 3.25. The experimental results show that the robot generates the elbow flexion motion when the lateral part of biceps is activated more than the lateral part of triceps. On the other hand, activation of the lateral part of triceps over the lateral part of biceps results the elbow extension motion. Moreover, it is also evident from the experimental results that the supination and pronation motion of the proposed inner skeleton robot can be controlled according to the EMG activation patterns of the lateral part of biceps and pronator teres, respectively. The experimental results depicted in Fig. 3.24 and 3.25 show that the proposed system generates the flexion-extension and supination-pronation motion according to the EMG signals of the muscles. Consequently, it is evident that the 2 DOF inner skeleton robot can be controlled according to the human muscles EMG signals and can assist a combination of elbow flexion-extension motion and forearm supination-pronation motion. Moreover, it is verified that the proposed T-mechanism keeps the similar configuration of the radius and ulna stem during forearm supination-pronation motion like the radius and ulna bone during normal human forearm motion.
**Figure 3.24**: Experimental Results for 2 DOF Inner Skeleton Robot – Slow Arm Motion.
FIGURE 3.25: Experimental Results for 2 DOF Inner Skeleton Robot – Fast Arm Motion.
DESIGN OF THE TRANSHUMERAL PROSTHESIS

The human above-elbow amputation can cause due to congenital (birth defect), trauma, tumor, disease, etc. out of which more than 80% of the amputation is caused due to trauma. A “Transhumeral Prosthesis” or “Above-Elbow Prosthetic Arm” is designed to compensate the lost functions of above-elbow (AE) amputees (who have lost their arm just above the elbow joint) absent arm. The ultimate purpose of the prosthetic arm is to provide an independent life of amputees as well as to provide more productive role of these people in the society. Recent progress in biomechatronics technology has facilitated increased mobility of AE amputees in performing daily life activities. A number of commercial prosthetic arm have been developed over the last few decades. However, many amputees have resisted using these prosthetics due to the high cost and the discrepancy between their expectations and reality. One of the main factors causing a lack of interest in current commercial prosthetic arms is the lack of desired DOF (degree of freedom) that results unnatural arm motion. In order to increase the mobility and to improve the quality of life of AE amputees in performing daily life activities, a 5-DOF externally powered prosthetic arm is proposed in this thesis. The proposed prosthesis is designed to generate natural human like arm motion during performing daily life activities. This chapter summarizes the mechanical design and the controllability of the proposed 5 DOF above-elbow prosthetic arm.

4.1 5 DOF TRANSHUMERAL PROSTHESIS

The main concerns of the amputees regarding their prosthesis are aesthetics, discomfort, excessive weight, poor functional capabilities, low controllability,
problems with technical assistance, noise and problems with the skin of the stump arm (irritation, sweating, etc.) [102] [103]. Currently, available commercial expensive cosmetic prostheses offer a more natural appearance and simple control. However, prosthesis available on the market for AE amputees provides a limited DOF. Most of these prostheses provide elbow flexion-extension motion with a terminal device attached at the end. In addition to the elbow motion, some prostheses provide forearm supination-pronation motion and a single DOF at the terminal device for grasping objects. Some passive DOF, which are useful to generate an optimal pre-determined configuration during performing certain tasks [60], are sometimes included in the prostheses. However, their dexterity is relatively very poor compared to the human arm. The human arm provides 7 DOF motion and, generates precise and complex motions during daily life activities which are almost impossible to be generated by using a limited DOF prosthetic arm. As a result, the presently available commercial prostheses have failed to gain wide acceptance among AE amputees.

In order to improve the quality of life of AE amputees and to increase their mobility in daily life activities (like, eating, drinking, dressing, brushing etc.), a 5-DOF externally powered transhumeral prosthesis is developed. The proposed prosthesis is designed to generate elbow flexion-extension, forearm supination-pronation, wrist flexion-extension and radial-ulnar deviation, and hand cylindrical grasp-release motion. Currently, no commercial transhumeral prosthesis provides a combination of wrist flexion-extension and radial-ulnar deviation motion, which have uttermost importance in daily life activities. Human wrist motion in combination with the forearm and upper arm motions control the position and rotation of the hand and allow us to perform daily activities such as feeding oneself, drinking a glass of water, turning a key in a lock, performing personal hygiene, etc. [44] [45]. The proposed 5 DOF transhumeral prosthesis for AE amputees is shown in Fig. 4.1. The prosthesis is designed according to the average size of a human adult male arm. The elbow-wrist length (the distance between the elbow joint and the distal part of the radius) is 280 mm, and the elbow-hand length (the distance between the elbow joint and the tip of the middle finger) is 450 mm. The proposed prosthesis is designed to provide sufficient joint torque compare to the normal human arm. The average total weight of the human forearm and hand is about 2 Kg and the weight of the prosthesis is kept about 2.3 Kg.
To imitate human arm function during prosthesis design, it is essential to use flexible actuators that can act like human muscles. However, electroactive polymer, Mckibben muscles, and shape memory alloy wires have not developed enough yet to use as an actuator for the prosthetic devices. In this study, five conventional rotary DC electric motors along with gear heads are used to generate the required 5 DOF motion. The actuation system of the prosthetic arm is listed in Table 4.1. In the human arm, muscles (act as actuators) for elbow motion carry the load of forearm and hand during elbow movement. The forearm consist the muscles which generate forearm, wrist and hand grasp-release motion. However, each of these muscles does not carry the load of others during the participation in corresponding joint motion. The similar principle is maintained to place the actuators in the proposed prosthesis to mimic the human anatomy. In the proposed prosthesis, the elbow motor carries the load of the remaining 4 actuators. These 4 actuators carry their own load while generates the corresponding motion. It can be noted that, in the human arm the muscles for elbow
motion lies in the upper arm. However, in the proposed prosthesis, the elbow motor is placed on the proximal part of the forearm to leave sufficient space for the stump arm to fit into the socket (i.e., stump arm holder) of the prosthesis. The forearm actuator, two wrist actuators, and also the actuator for hand grasp-release motion, are placed in the forearm part of the prosthesis. Placing all of the five actuators in the proximal part of the prosthesis forearm not only reduces the inertia effect during prosthesis movement, but also provides sufficient grasping space in the palm. The motors and the gearheads are selected to afford sufficient amount of joint torque for daily life activities.

**Table 4.1: Actuation System for 5 DOF Prosthetic Arm**

<table>
<thead>
<tr>
<th>Joint</th>
<th>Motor</th>
<th>Gearhead</th>
<th>Encoder</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Elbow F-E Motion</strong></td>
<td>Nominal Voltage 24 v</td>
<td>Double Shaft Type</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Nominal Speed 8050 rpm</td>
<td>Reduction 100 :1</td>
<td>Line Driver</td>
</tr>
<tr>
<td></td>
<td>Nominal Torque 85 mNm</td>
<td>Rated Torque 2.4 Nm</td>
<td>Resolution 1000</td>
</tr>
<tr>
<td></td>
<td>Nominal Current 3.44 A</td>
<td>Peak Torque 4.8 Nm</td>
<td>Channel 3</td>
</tr>
<tr>
<td></td>
<td>Starting Current 39.3A</td>
<td>Max. Torque 9 Nm</td>
<td>Voltage 5v</td>
</tr>
<tr>
<td></td>
<td>Efficiency 87%</td>
<td>Weight 120 gm</td>
<td>Current 5mA</td>
</tr>
<tr>
<td></td>
<td>Weight 238 gm</td>
<td>Max. I/P 8500 rpm</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Overall Length (Motor + Encoder) – 100 mm</td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Forearm S-P Motion</strong></td>
<td>Nominal Voltage 24 v</td>
<td>Planetary Gearhead</td>
<td></td>
</tr>
<tr>
<td><strong>Wrist F-E Motion</strong></td>
<td>Nominal Speed 8360 rpm</td>
<td>Reduction 84:1</td>
<td>Line Driver</td>
</tr>
<tr>
<td><strong>Wrist R-U Motion</strong></td>
<td>Nominal Torque 26.7 mNm</td>
<td>Max. Torque 1.3 Nm</td>
<td>Resolution 1000</td>
</tr>
<tr>
<td></td>
<td>Nominal Current 1.17 A</td>
<td>Weight 108 gm</td>
<td>Channel 3</td>
</tr>
<tr>
<td></td>
<td>Starting Current 11A</td>
<td>Max. Efficiency 59%</td>
<td>Voltage 5v</td>
</tr>
<tr>
<td></td>
<td>Efficiency 86%</td>
<td>O/P Shaft Stainless Steel</td>
<td>Current 5mA</td>
</tr>
<tr>
<td></td>
<td>Weight 130 gm</td>
<td>O/P Bearing Ball bearing</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Overall Length (Motor + Gearhead + Encoder) – 108.4 mm</td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Hand G-R Motion</strong></td>
<td>Nominal Voltage 12 v</td>
<td>Reduction 80:1</td>
<td>Open Collector</td>
</tr>
<tr>
<td></td>
<td>Rated Speed 55 rpm</td>
<td>Resolution 360</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Peak Speed 110 rpm</td>
<td>Channel 2</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Rated Torque 0.29 Nm</td>
<td>Voltage 5v</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Peak Torque 0.59 Nm</td>
<td>Current 5mA</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Overall – Length : 89 mm, Mass : 90 gm</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
4.3 **Movable Range of Each Joint of the Prosthetic Arm**

The human arm provides 7 DOF motion and allows different range of motion at its articulations. However, to perform daily life activities, the required range of motion is usually less than the maximum range of motion provided by the human arm articulations. The human elbow complex allows two DOF, elbow flexion-extension motion and forearm supination-pronation motion. Usually, the limitation of the movable range of the elbow flexion-extension motion is 145 degrees in flexion and 5 degrees in extension and that of the forearm supination-pronation motion is 80-90 degrees in supination and 50-80 degrees in pronation. The proposed prosthetic arm is designed to generate 0°-140° of elbow flexion-extension motion, and 0°-80° of forearm supination-pronation motion. In the human arm, the maximum movable range of wrist motion is 65°-85° of flexion, 50°-70° of extension, 15°-25° of radial deviation, and 25°-45° of ulnar deviation. For daily life activities, the functional range of wrist motion is 5° of flexion, 30° of extension, 10° of radial deviation, and 15° of ulnar deviation [105]. The prosthetic arm is designed to allow 50° for both of wrist flexion and extension motion, and that of for radial & ulnar deviation is kept 30° for both. The human hand provides several grasp (spherical, cylindrical, lateral, tin pinch, etc.) and gesture abilities. The normal range of motion for human hand open-close motion is 0°-90° [106]. In this study, we consider only the hand cylindrical open-close (i.e., grasp-release) motion, and the prosthesis hand is designed to provide 0°-70° range of motion. The comparison of the range of motion (ROM) between the human arm and the proposed prosthetic arm is listed in Table 4.2.

**Table 4.2: ROM for the Human Arm and the Proposed Prosthetic Arm**

<table>
<thead>
<tr>
<th>Motion</th>
<th>Human Arm (Maximum ROM)</th>
<th>Prosthetic Arm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Elbow F-E Motion</td>
<td>145°- Flex. &amp; 5° - Ext.</td>
<td>140° - Flex. &amp; 0° - Ext.</td>
</tr>
<tr>
<td>Forearm S-P Motion</td>
<td>80°-90° - Sup. &amp; 50°-80° - Pro.</td>
<td>0°-80° Supination-Pronation</td>
</tr>
<tr>
<td>Wrist F-E Motion</td>
<td>65°-85° - Flex. &amp; 50°-70° - Ext.</td>
<td>50° - Flex. &amp; 50° - Ext.</td>
</tr>
<tr>
<td>Hand G-R Motion</td>
<td>0°-90° Grasp-Release</td>
<td>0°-70° Grasp-Release</td>
</tr>
</tbody>
</table>
4.4 **ELBOW MOTION**

![Diagram of Elbow Joint Mechanism of the Prosthetic Arm](image)

**Figure 4.2:** Elbow Joint Mechanism of the Prosthetic Arm.

The elbow joint of the proposed prosthesis which is shown in Fig. 4.2, provides 4.8 Nm joint torque at maximum and permits 120° flexion and 0° extension motion. One of the main features of the designed prosthesis is that the elbow joint actuator is placed on the proximal part of the forearm which provides sufficient space in the socket (i.e., stump arm holder) to hold user’s stump arm as the level of amputation differs from person to person. In order to retain the breadth size of the proximal part of forearm, the gearhead (Harmonic Drive, CSF-8-100-1U) is used in parallel with the elbow motor (Maxon Motor, RE 30, Graphite Brushes, 60 Watt). The proximal part of the prosthetic arm is mainly consist a base named as “elbow base” that holds the five actuators used for the prosthetic arm. The elbow motor and the gearhead are placed in parallel in the elbow base as shown in Fig. 4.2. The position to fix the remaining 4 actuators on the elbow base is also shown in Fig. 4.2. The elbow motor torque is transferred to the gearhead input shaft by means of a pair of spur gears and the gearhead output shaft is also equipped with a spur gear. The stump arm holder holds a main supporting shaft (ends are firmly attached with the socket) that connects the socket with the forearm part. Two bearing holders with bearings are used in the proximal forearm part so that the supporting shaft can be placed inside of the two bearings. A spur gear is rigidly fixed with the supporting shaft in such a way that its
teeth mate with the gearhead output shaft gear teeth. The designed elbow joint mechanism ensures that the rotation of the elbow motor will cause the whole forearm part to rotate relative to the stump arm holder part, like the human elbow flexion-extension motion. The mechanical design of the parts used for elbow motion is shown in Figs. 4.3.

**FIGURE 4.3.1:** Parts of the Elbow Mechanism.

**FIGURE 4.3.2.1:** Design of the Elbow Base (EP-1).
FIGURE 4.3.2.2: Design of the Elbow Base (EP-1).

FIGURE 4.3.3: Elbow Motor Holder (EP-2).
FIGURE 4.3.4: Gearhead I/P Holder (EP-3).

FIGURE 4.3.5: Gearhead O/P Holder (EP-4).

FIGURE 4.3.6: Support Shaft (EP-5).
4.5 FOREARM MOTION

The designed prosthesis provides 0°-70° forearm supination-pronation motion and permits 1.32 Nm joint torque at maximum. In the existing externally powered above-elbow prosthetic arm, the terminal device (to grasp objects) is usually mounted on a circular rotating element to provide forearm supination-pronation motion, which is different from their biological counterpart. The human forearm consist two bones; radius and ulna, and the movement of these two bones generates the forearm motion. In the human arm, the radius bone of the forearm crosses over the ulna bone during pronation motion whereas during supination motion they positioned almost parallel to each other. The position of the radius and ulna bone during supination-pronation motion has already shown in Fig. 2.6. In order to mimic the human-like forearm motion, a modified T-Mechanism is introduced here for the transhumeral prosthesis (i.e., above-elbow prosthetic arm) design. A schematic diagram of the T-mechanism for above-elbow prosthetic arm is shown in Fig. 4.4. The T-mechanism is formed by the combination of four ball joints, a T-link, the shaft of the forearm actuator, and two steel shafts. Two shafts are used like radius and ulna bone in the prosthesis forearm which also connects the wrist part with the forearm. A schematic diagram of the forearm motion mechanism is shown in Fig. 4.5. The radius and ulna shafts are placed
between proximal and distal part of the forearm (wrist) by means of four ball joints. Two ball joints are placed inside of the two grooves at the elbow base and are firmly fixed there by using a ball joint holder. Similarly, other two ball joints are positioned at the wrist joint. The T-link as shown in Fig. 4.5 consists of two holes at its two ends and a base. The base of the T-link consist a hole inside of which the shaft of the forearm actuator is firmly fixed (Maxon Motor, RE 25, Graphite Brushes, 20 Watt + Planetary GH, GP26B, 0.5-2.0 Nm). The radius and the ulna shaft are allowed to pass through the end holes of the T-link. It can be noted that the holes size of the T-link have to kept a bit larger than the shaft (radius and ulna) diameter in order to accomplish the rotation of the radius over the ulna during pronation motion. The
The desired human-like forearm supination-pronation motion is acquired by transferring the forearm actuator torque to radius and ulna shaft through the T-link as well as the T-mechanism. The mechanical design of the parts used for forearm motion is depicted in Figs. 4.6.

**FIGURE 4.6.1**: Parts for Forearm Motion.

**FIGURE 4.6.2**: Ball Joint Holder (FP-1).

**FIGURE 4.6.3**: Forearm Actuator Holder (FP-2).
**FIGURE 4.6.4**: Forearm Actuator Holder (FP-3).

**FIGURE 4.6.5**: T – Link (FP-4).

**FIGURE 4.6.6**: Radius & Ulna (FP-5).

**FIGURE 4.6**: Mechanical Design of the Parts for Forearm Motion.
4.6 2 DOF WRIST MOTION

![Diagram of Wrist Actuators in Prosthesis]

**FIGURE 4.7**: Position of the Wrist Actuators in the Prosthesis.

The wrist joint of the proposed prosthesis permits two DOF motions, and the allowable range of motion is 50° for both of flexion and extension motion and that of for radial and ulnar deviation is kept 30° for both. Two actuators with 1.32 Nm continuous torque ratings are used to generate the 2 DOF wrist motions. The position of the actuators in the prosthetic arm is shown in Fig. 4.7. As stated earlier, during forearm motion, the radius crosses over the ulna. However, the ulna never crosses over the radius during forearm motion. This fundamental concept is implemented to position the wrist actuators inside of the forearm of the prosthesis. The actuator that provides wrist flexion-extension motion is placed on top of the ulna and the radial-ulnar deviation actuator is positioned on bottom of the radius by using two sets of actuator holders. The actuator holders are firmly fixed with the elbow base by using screws. It can be noted that the wrist actuator holders also act as a mechanical stopper during forearm motion. In order to retain the forearm breadth narrower toward the distal part like the human arm, the wrist actuators are placed at an inclination of about 5°. The torque of the actuators is transferred to the wrist joint by means of pulley and
Bowden cable mechanism. Two pulleys as shown in Fig. 4.7 are fixed with the output shaft of the actuators. The Bowden cable mechanism is formed by the combination of the hollow cable housings, inner cables, pulley and the wrist joint plates. In this study, 4 mm flexible helical steel wire coated with soft plastic are used as the cable housing. As shown in Fig. 4.7, the ends of the cable housing are fixed at the actuator holder and the wrist plate-1 by means of screws. The inner cables are passed inside of the cable housing and their ends are fixed with the pulley and the wrist plate-2. As the actuator output shaft rotates, the tangential force of the pulley is transmitted to the wrist joint by the movement of the 0.45 mm flexible steel inner cables relative to the hollow outer cable housing. The movements of the inner cables are used to generate pulling force at the wrist joint to provide the 2 DOF motions.

**FIGURE 4.8:** Direction of Pulling Forces for 2 DOF Wrist Motion.

The wrist joint is mainly consists of two wrist plates, a connecting rod and a ball joint which is shown in Fig. 4.8. The wrist plate-1 consist two grooves inside of which two ball joints are placed and positioned by using a ball joint support. The distal part of radius and ulna shaft are fixed inside of these two ball joints, and the radius and the ulna supports are used to prevent the wrist plate-1 motion due to the gravity force. The wrist plate-2 is mainly consists the hand parts and there is also a grove inside of which another ball joint is placed. The connecting rod between the two plates is firmly fixed at wrist plate-1 and is connected with plate-2 by the ball joint. The cables
housings for wrist motions are fixed with the plate-1 and the inner cables are passed across the plate-1 and firmly fixed with the plate-2. In order to provide a combination of two DOF motion, each pair of inner cables for each DOF are fixed on the same axis relative to the center of the ball joint (placed at the plate-2). Since the connecting rod is fixed with the plate-1 and attached with the plate-2 by means of a ball joint, pulling the inner cables generates the required wrist motion which is shown in Fig. 4.8. The length of the connecting rod limits the range of wrist motions and the plate-1 also serves as a mechanical stopper. It can be noted that as the inner cables are attached with the pulley, pulling one end of the cable will release the opposite end. The mechanical design of the parts used for the wrist motion is shown in Figs. 4.9.

**FIGURE 4.9.1:** Parts for Wrist Motion.

**FIGURE 4.9.2.1:** Wrist Plate-1 (WP-1).
FIGURE 4.9.2.2: Wrist Plate-1 (WP-1).

FIGURE 4.9.3.1: Wrist Plate-2 (WP-2).

FIGURE 4.9.3.2: Wrist Plate-2 (WP-2).
FIGURE 4.9.4.1: Ball Joint Holder (WP-3).

FIGURE 4.9.4.2: Ball Joint Holder (WP-3).

FIGURE 4.9.5: Radius Support (WP-4).
**FIGURE 4.9.6**: Pulley (WP-5).

**FIGURE 4.9.7.1**: Actuator Holder (WP-6).

**FIGURE 4.9.7.2**: Actuator Holder (WP-6).
4.7 **HAND MOTION**

The hand of the proposed prosthesis is designed to provide the cylindrical grasp-release motion, which is important for the daily life activities such as, drinking a glass of water, holding a bottle, grasping a can, etc. In order to provide sufficient grasping space in the palm and to reduce the inertia effect, the hand actuator (Harmonic Drive, RH-5A-5502-036AO) is placed in the forearm part of the prosthesis. The position of the hand actuator in the prosthesis and the prosthesis hand are shown in Figs. 4.10 and
4.11, respectively. The palm of the proposed prosthesis hand is defined by two metal plates inside which two shafts are placed to hold the fingers and the thumb by using four ball bearings. The four fingers of the prosthesis hand are located on the outermost edge of the palm and the thumb is located on one side of the palm. The cylindrical grasp-release motion in the prosthesis hand can be realized by generating two opposite direction of motion on the fingers and the thumb. In order to generate opposite direction of motion between the fingers and the thumb, two shafts are connected by using a pair of spur gears. The torque of the hand actuator is transmitted to the hand by using pulleys and Bowden cable mechanism. Two pulleys, two flexible inner cables (0.45 mm steel cable) and two cable housing are used to transmit the actuator power to the hand. The two pulleys (similar in dimension with that used for
wrist actuators) are mounted on the output shaft of the hand actuator and the thumb shaft, and are interconnected by the inner cables. The ends of the outer cables are fixed at the actuator holder (HP-6) and wrist plate-2. The movement of the inner cables inside of the hollow cable housings transmits the motor power to the thumb shaft. Thus the rotation of the hand actuator generates the opposite direction of motion on prosthesis fingers and thumb, as well as generates the grasp-release motion. It can be noted that the hand cable housings can move freely inside of the wrist plate-1 during wrist flexion-extension motion. The mechanical design of the prosthesis hand parts are depicted in Figs. 4.12.

**FIGURE 4.12.1**: Parts for Hand Motion.

**FIGURE 4.12.2**: Thumb (HP-1).
FIGURE 4.12.3: Upper Plate (HP-2).

FIGURE 4.12.4: Finger (HP-3).

FIGURE 4.12.5: Finger and Thumb Holder (HP-4).
FIGURE 4.12.6: Lower Plate (HP-5).

FIGURE 4.12.7.1: Actuator Holder (HP-6).

FIGURE 4.12.7.2: Actuator Holder (HP-6).

FIGURE 4.12: Mechanical design of the Hand Motion Parts.
4.8 EXPERIMENTS

Experiments have been carried out to evaluate the controllability of the proposed prosthesis. In order to acquire the angular position feedback of each joint, all of the actuators are equipped with an optical position encoder. The joint angle is measured using the encoder signal and the prosthesis dimension. In order to verify the controllability, the stump arm holder of the prosthesis is designed as a rectangular shape. The experimental set-up as shown in Fig. 4.13 consists of a personal computer, an interface board (RITECH, RIF-171-1), five motor drivers (iXs Research Corp., iMDs03), a DC power supply, and the proposed prosthesis. In this experiment, Proportional-Derivative (PD) controller has been implemented to evaluate the controllability of the prosthesis motions and the sinusoidal motion trajectory has been used as the desired trajectory. The amplitude of the desired trajectory was selected according to the allowable range of motion of the corresponding joint. The feedback torque command ($\tau_m$) of the motors is computed as,

$$\tau_m = K_p (\theta_d - \theta) + K_v (\dot{\theta}_d - \dot{\theta}) \tag{4.1}$$

Here, $K_p$ and $K_v$ are the position and velocity gain, respectively. $\theta_d = [\theta_{d1}, \ldots, \theta_{dn}]^T$ and $\theta = [\theta_1, \ldots, \theta_N]^T$ are the desired and the measured joint angle vector, respectively. $N$ is the number of total samples. The torque command is applied to the corresponding motor via the motor driver. The experimental results for elbow flexion-extension, forearm supination-pronation, wrist flexion-extension, radial-ulnar deviation, and hand cylindrical grasp-release motions are shown in Figs. 4.14-4.18, respectively. For each of the motion, two experimental results with different range of
motion are depicted. As mentioned earlier, the maximum range of wrist motion can be varied by changing the length of the connecting rod that exists between the wrist plates. In this study, the controllability of the 2 DOF wrist motion with different range of motion is verified by changing the connecting rod length. The experimental results show that the proposed prosthesis can generate the desired range of motion and can follow the desired trajectory.

**Figure 4.14**: Experimental Results – Elbow Flexion-Extension Motion.

**Figure 4.15**: Experimental Results – Forearm Supination-Pronation Motion.
FIGURE 4.16: Experimental Results – Wrist Flexion-Extension Motion.

FIGURE 4.17: Experimental Results – Wrist Radial-Ulnar Deviation Motion.
FIGURE 4.18: Experimental Results – Hand Grasp-Release Motion.
Control of a multi degree-of-freedom (DOF) prosthetic arm remains as a challenging problem. The poor controllability of existing prosthetic arms is one of the main reasons causing a lack of interest of using these prostheses. Since the invention of the externally powered prosthetic arm, a number of controller techniques have already been proposed for the control of prosthetic arm. In myoelectric prosthesis controller technique, user’s stump arm or other body part muscles surface electromyogram (EMG) signals are used as input command signals to actuate the electromechanical active joints of the prosthetic arm. Recently, targeted muscle reinnervation technique was revealed to improve the myoelectric prosthesis control. In this technique, the active nerves from the stump arm are transferred to the other body part muscles and the EMG signals of those muscles are used to control the prosthetic arm. In muscle tunnel cineplasty technique, a part of the stump muscles are fitted with the prosthetic arm by surgical operation to control the active joint of the prosthetic arm. Control of a prosthetic arm using the kinematic data of the remaining normal arm was also proposed few years ago. Another widely known approach for control of a prosthetic arm is the extended physiological proprioception (EPP) technique that was proposed in 1972 by D. C. Simpson. In this technique, residual or stump arm joint kinematics are used as controller input information to control the prosthetic arm. Although there are numerous controller methods have already been proposed for the control of the prosthetic arm, there is no exaggeration to mention that the myoelectric prosthesis controller technique is extensively accepted among these concepts. Nowadays, the most advanced commercially available externally powered prosthetic arm is the myoelectric prosthetic arm. However, currently available
myoelectric prosthetic arm provides a limited DOF motion and it is not easy to control the multi DOF prosthetic arm using the myoelectric prosthesis controller technique. Moreover, there is no evidence still exist that shows the previously mentioned prosthetic arm controller technique can be effectively implemented for multi DOF prosthetic arm as well as for the 5 DOF prosthetic arm developed in this study. As a result, a controller technique is proposed in this study for the control of the 5 DOF above-elbow prosthetic arm. This chapter summarizes the proposed controller technique.

5.1 PROPOSED CONTROL STRATEGY FOR 5 DOF PROSTHETIC ARM

The human arm biceps and triceps muscles are responsible to generate elbow flexion-extension motion and located in the upper arm. Among above-elbow (AE) amputees, the arm amputation is usually found at just above the elbow joint and the stump upper arm part consist biceps and triceps muscles. As a result, the proposed prosthetic arm controller is designed based upon the assumption that user’s biceps and triceps muscles are remained and can be activated according to the user’s intension. The prosthetic arm is supposed to be controlled by using a combination of the EMG signals based controller (EBC) and the task oriented kinematics based controller (KBC). User’s stump arm muscles EMG signals and stump arm joint kinematics (angular position, angular velocity and angular acceleration) are used together to control the 5 DOF motion of the prosthetic arm. The EMG signals of the amputee’s biceps and triceps muscles are used as input information for the EBC to control prosthesis elbow flexion-extension and hand grasp-release motion. In order to control forearm supination-pronation, wrist flexion-extension, and radial-ulnar deviation motions of the prosthetic arm, at the first step, a task (i.e., activity) classifier is used to identify amputee’s intended activity by using amputee’s stump arm shoulder joint kinematics and prosthesis elbow kinematics. After identification of the user’s intended activity, the desired hand trajectory with respect to a fixed shoulder coordinate system are estimated based on the nature of the classified task in the second step. In the third step, the KBC is used for the control of prosthesis forearm and wrist motions to realize the objective of the amputee’s intended activity. A schematic diagram of the proposed control strategy is shown in Fig. 5.1.
5.2 EMG Signals Based Controller (EBC)

The human muscles electromyogram (EMG) signals are among the most important biological signals that directly reflects the human motion intension. As a result, most of the currently available prosthetic arms used EMG signals as controller input information. Muscles EMG signals is usually stochastic (i.e., random) in nature and the maximum amplitude of EMG signals can be varied time to time depending upon the physical condition of human. Commercially available myoelectric AE prosthetic arms use the muscles EMG signals as switches for on/off control of the actuators (i.e., electromechanical active joints) which actuate the prosthetic arm segments. Therefore, amputee’s motion intension can not be precisely reflected in the generated arm motion. In this study, an EMG signals based fuzzy controller (i.e., EBC) is designed that controls the torque of elbow and hand actuators of the prosthetic arm proportionally to the amount of EMG signals. Thus the EMG signals are not used as simple switches in this controller.

The surface EMG signals of the user’s short head of biceps (ch.1), long head of biceps (ch.2), lateral head of triceps (ch.3), and long head of triceps (ch.4) muscles, are used as input information for the EBC. The position of the surface electrodes to measure EMG signals from the designated muscles is shown in Fig.5.2. In this controller technique, elbow flexion motion is generated when biceps are activated and activation of the triceps muscles results the elbow extension motion. Hand grasp-
release motion is controlled using the short head of biceps and the long head of triceps. Hand grasp motion is generated when both of biceps and triceps muscles are activated simultaneously. The hand remains in release position when both of biceps and triceps are not working. Since it is difficult to use raw EMG signals for input information, root mean square (RMS) value is calculated to extract the feature of the EMG signals. The RMS is determined as,

$$RMS = \sqrt{\frac{1}{N} \sum_{i=1}^{N} v_i^2}$$  \hspace{1cm} (5.1)

where, $v_i$ is the voltage value at $i^{th}$ sampling and $N$ is the number of samples in a segment. The input variables for the EBC are the four RMS values of the EMG signals. Three kinds of fuzzy linguistic variables (Zero: ZO, Positive Small: PS and Positive Big: PB) are prepared for each input which are shown in Fig. 5.3. Two kinds of nonlinear functions ($f_G$ and $f_S$) are used to express the membership function of the fuzzy controller:

$$f_s(u_s) = \frac{1}{1 + e^{-u_s}}$$  \hspace{1cm} (5.2)

$$u_s(x) = \omega_o + \omega_s x$$  \hspace{1cm} (5.3)

$$f_G(u_G) = e^{-v_G^2}$$  \hspace{1cm} (5.4)

$$u_G(x) = \frac{\omega_o + x}{\omega_i}$$  \hspace{1cm} (5.5)

where, $x$ is the input signal, $\omega_o$ is a threshold value and $\omega_i$ is a weight.
The weight value, $w_i$ and threshold value, $w_o$ for the ZO membership function are calculated as follows:

$$w_i = \frac{\log((1.0 / \text{H\_RATE} - 1)/(1.0 / \text{L\_RATE} - 1)))}{(w_3 - w_1)}$$

(5.6)

$$w_o = ((-1.0) \log(1.0 / \text{H\_RATE} - 1.0) - w_i w_i)$$

(5.7)

For the PS membership function, $w_i$ and $w_o$ are calculated as:

$$w_i = \sqrt{(-w_3 + (w_3 + w_3) / 2)^2 / \log(1.0 / \text{M\_RATE})}$$

(5.8)

$$w_o = -w_3$$

(5.9)

For the PB membership function, $w_i$ and $w_o$ are calculated as:

$$w_i = \frac{\log((1.0 / \text{H\_RATE} - 1)/(1.0 / \text{L\_RATE} - 1)))}{(w_3 - w_3)}$$

(5.10)

$$w_o = ((-1.0) \log(1.0 / \text{H\_RATE} - 1.0) - w_i w_3)$$

(5.11)

In this study, ten kinds of fuzzy IF-THEN control rules are prepared to control the prosthesis elbow joint torque and seven IF-THEN control rules are made to control the hand torque. The fuzzy IF-THEN control rules for elbow and hand are listed in Tables 5.1 and 5.2, respectively. The output of the fuzzy controller is calculated using the following equation:
\[ O = \frac{\sum_{i} \omega_{ni} y_{ki}}{\sum_{i} y_{ki}} \]  \hspace{1cm} (5.12)

where,

- \( O \) = Output vector;
- \( y_{ki} \) = Degree of fitness of \( i_{th} \) rule;
- \( \omega_{ni} \) = Weight for \( i_{th} \) rule;

**Table 5.1: Fuzzy IF-THEN Control Rules for Elbow Flexion-Extension Motion**

<table>
<thead>
<tr>
<th>Rule</th>
<th>IF</th>
<th>THEN</th>
</tr>
</thead>
<tbody>
<tr>
<td>01</td>
<td>EMG ch.1 is PS and EMG ch.2 is PS</td>
<td>Actuator Torque = 1.25 Nm</td>
</tr>
<tr>
<td>02</td>
<td>EMG ch.1 is PB and EMG ch.2 is PB</td>
<td>Actuator Torque = 4.0 Nm</td>
</tr>
<tr>
<td>03</td>
<td>EMG ch.1 is PS and EMG ch.2 is PB</td>
<td>Actuator Torque = 2.5 Nm</td>
</tr>
<tr>
<td>04</td>
<td>EMG ch.1 is PB and EMG ch.2 is PS</td>
<td>Actuator Torque = 2.5 Nm</td>
</tr>
<tr>
<td>05</td>
<td>EMG ch.3 is PS and EMG ch.4 is PS</td>
<td>Actuator Torque = -1.25 Nm</td>
</tr>
<tr>
<td>06</td>
<td>EMG ch.3 is PB and EMG ch.4 is PB</td>
<td>Actuator Torque = -4.0 Nm</td>
</tr>
<tr>
<td>07</td>
<td>EMG ch.3 is PS and EMG ch.4 is PB</td>
<td>Actuator Torque = -2.5 Nm</td>
</tr>
<tr>
<td>08</td>
<td>EMG ch.3 is PB and EMG ch.4 is PS</td>
<td>Actuator Torque = -2.5 Nm</td>
</tr>
<tr>
<td>09</td>
<td>EMG ch.1 is PS and EMG ch.2 is ZO and EMG ch.3 is ZO</td>
<td>Actuator Torque = 0 Nm</td>
</tr>
<tr>
<td>10</td>
<td>EMG ch.1 is PB and EMG ch.2 is ZO and EMG ch.3 is ZO</td>
<td>Actuator Torque = 0 Nm</td>
</tr>
</tbody>
</table>

**Table 5.2: Fuzzy IF-THEN Control Rules for Hand Grasp-Release Motion**

<table>
<thead>
<tr>
<th>Rule</th>
<th>IF</th>
<th>THEN</th>
</tr>
</thead>
<tbody>
<tr>
<td>01</td>
<td>EMG ch.1 is PB and EMG ch.4 is PB</td>
<td>Actuator Torque = 0.3 Nm</td>
</tr>
<tr>
<td>02</td>
<td>EMG ch.1 is PS and EMG ch.4 is PB</td>
<td>Actuator Torque = 0.24 Nm</td>
</tr>
<tr>
<td>03</td>
<td>EMG ch.1 is PB and EMG ch.4 is PS</td>
<td>Actuator Torque = 0.24 Nm</td>
</tr>
<tr>
<td>04</td>
<td>EMG ch.1 is PS and EMG ch.4 is PS</td>
<td>Actuator Torque = 0.21 Nm</td>
</tr>
<tr>
<td>05</td>
<td>EMG ch.1 is ZO and EMG ch.4 is ZO</td>
<td>Actuator Torque = -0.3 Nm</td>
</tr>
<tr>
<td>06</td>
<td>EMG ch.1 is PS and EMG ch.4 is ZO</td>
<td>Actuator Torque = -0.21 Nm</td>
</tr>
<tr>
<td>07</td>
<td>EMG ch.1 is ZO and EMG ch.4 is PS</td>
<td>Actuator Torque = -0.21 Nm</td>
</tr>
</tbody>
</table>
5.3 Classification of the Daily Life Activities – Artificial Neural Network Based Task Classifier

To generate the desired wrist and forearm motions for the prosthetic arm, amputee’s intended activity is identified in the first step. In order to identify the intended activity, a task classifier that classifies the daily activities (i.e., tasks) using amputee’s stump arm shoulder and prosthesis elbow kinematics is designed. In this study, 10 daily life activities that are important and frequently performed in daily living are considered. To be used for the control of the prosthetic arm, the task classification must be carried out precisely and quickly. Moreover, it is expected that the classifier classifies the tasks in real time. It is inevitable that the amputees will become frustrated if the prosthetic arm takes a much longer delays between the input signals and the activation of the prosthetic arm. Furthermore, the classifier must be able to classify the tasks with the data which are not exactly identical with previously learned data. In other words, the classifier should be adaptable for different users. In order to obtain the above mentioned requirements, a multilayer artificial neural network (ANN) has been designed as a classifier to classify the daily life activities. It is known that the multilayer neural network is suitable for classification problem such as pattern recognition [106]-[109]. The architecture of the ANN based classifier is shown in Fig. 5.4. It consists of three layers: input layer, hidden layer (HL), and output layer. There are 12 neurons in the input layer, 200 neurons in the HL and 10 neurons in the output layer. The kinematics ($\theta$, $\dot{\theta}$, and $\ddot{\theta}$) of the shoulder flexion-extension, abduction-adduction, internal-external rotation, and elbow flexion-extension motion, are used as the input signals for the classifier. Since there is no established rule to define the number of HL neurons, a number of trials have been performed with different number of HL neurons to define the number of HL neurons. The number of output layer neurons have decided based upon the number of selected daily life activity (one neuron for each activity). The error back propagation learning algorithm has been applied to train the classifier. Nonlinear unipolar sigmoid function ($f_S$) is chosen as the activation function for the neurons in HL and output layer as their derivatives are easy to calculate. It is helpful for calculating the weight values during error back propagation learning algorithm. The activation or transfer function for a unipolar sigmoid function can be written as:
FIGURE 5.4: Artificial Neural Network based Task Classifier.

\[ a(f) = \frac{1}{1 + e^{-\lambda f}} \]  

where, \( \lambda \) is the steepness parameter for the sigmoid function. As shown in Fig. 7, \( w_{ij} \) (\( N_2 \times N_1 \) matrix) represents the weight value between the input layer and HL and \( v_{jk} \) (\( N_3 \times N_2 \) matrix) is the weight value between the HL and output layer. The output signals from the output layer is supposed to be “1” for the classified activity-\( N \): \( O_N \) and “0” for the others.

5.4 Desired Trajectory of the Hand

The daily life activities selected in this study are listed in Table 5.3 and designated as Task1-Task10 here. For a particular task, human hand trajectory with respect to a fixed shoulder coordinate system can be estimated. The desired hand trajectory to fulfill the objective of a task can be estimated based on the nature of the task. Figure 5.5 shows the required hand orientation to execute four different tasks. In order to define the desired hand trajectory to realize a task, selected tasks are defined into either two or three phase (i.e., state of the task) based on the nature of the task. As shown in Fig. 5.5(b) and (c), hand trajectory for task-1 and task-7 can be shown by using initial and final orientation of the hand. However, in case of task-4 and task-5,
TABLE 5.3: SELECTED DAILY LIFE ACTIVITIES

<table>
<thead>
<tr>
<th>Task-1</th>
<th>Reach and Grasp an Object</th>
<th>Task-6</th>
<th>Pouring Water from a Bottle</th>
</tr>
</thead>
<tbody>
<tr>
<td>Task-2</td>
<td>Eat With Spoon</td>
<td>Task-7</td>
<td>Pick a Phone to Ear Level</td>
</tr>
<tr>
<td>Task-3</td>
<td>Brush Teeth</td>
<td>Task-8</td>
<td>Shave</td>
</tr>
<tr>
<td>Task-4</td>
<td>Comb Hair</td>
<td>Task-9</td>
<td>Wash the Face</td>
</tr>
<tr>
<td>Task-5</td>
<td>Drink a Glass of Water</td>
<td>Task-10</td>
<td>Open a Drawer</td>
</tr>
</tbody>
</table>

FIGURE 5.5: Hand Orientation – (a) Base Coordinate System (b) Reach and Grasp an Object (c) Pick a Phone to Ear Level (d) Comb Hair and (e) Drink a Glass of Water.

which are shown in Figs. 5.5(d) and (e), respectively; after the hand reached at an intermediate phase, there is necessity of hand pronation motion to execute the intended task effectively. In order to generate the required hand orientation to accomplish the objective of the task, the desired wrist and forearm angles w.r.t. the shoulder and elbow angles are estimated for all kinds of selected tasks. As the orientation of the hand to fulfill the objective for all kinds of selected tasks are estimated, the trajectory of the hand for each task (from initial to final state) are calculated by using direct kinematics.
A simplified kinematic model of the 7 DOF human arm with the reference and link coordinate systems using the Denavit-Hartenberg (D-H) convention is shown in Fig. 5.6. This model considered a fixed coordinate system (0) at the shoulder and neglected the shoulder scapula movement. The D-H parameters can be defined as follows:

\[ a_i \]: Offset Distance between two adjacent axes.
\( d_i \): Translation Distance between two incident normals of a joint axis.

\( \alpha_i \): Twist angle between two adjacent joint axes. It is the angle required to rotate the \( Z_{i-1} \) axis into alignment with the \( Z_i \) axis about the positive \( X_i \) axis according to the right-hand rule.

\( \theta_i \): Joint angle between two incident normals of a joint axis. It is the angle required to rotate the \( X_{i-1} \) axis into alignment with the \( X_i \) axis about the positive \( Z_{i-1} \) axis according to the right hand rule.

Using the above parameters, the Denavit-Hartenberg (D-H) transformation matrix can be written as shown in Eq. (5.14).

\[
\begin{bmatrix}
    c\theta_i & -c\alpha_i s\theta_i & s\alpha_i s\theta_i & a_i c\theta_i \\
    s\theta_i & c\alpha_i c\theta_i & s\alpha_i c\theta_i & a_i s\theta_i \\
    0 & s\alpha_i & c\alpha_i & d_i \\
    0 & 0 & 0 & 1
\end{bmatrix}
\]

(5.14)

where, \( c\theta_\eta \) and \( s\theta_\eta \) are \( \cos \theta_\eta \) and \( \sin \theta_\eta \), respectively. The trailing subscript \( i \) and the leading subscript \( i-1 \) denote that the transformation takes place from the \( i^{th} \) coordinate system to the \( (i-1)^{th} \) coordinate system. For the coordinate systems chosen as shown in Fig. 5.6(b), the D-H parameters of the arm are listed in Table 5.4, where \( L_3 \), \( L_5 \), and \( L_7 \) are the length of the upper arm, forearm and hand, respectively. The transformation matrices between frame \( i \) (\( i = 1….7 \)) and \( i-1 \) is obtained by substituting the D-H parameters into the D-H transformation matrices which are shown in Eq. (5.15)-(5.21).

\[
\begin{bmatrix}
    c\theta_i & 0 & -s\theta_i & 0 \\
    s\theta_i & 0 & c\theta_i & 0 \\
    0 & -1 & 0 & 0 \\
    0 & 0 & 0 & 1
\end{bmatrix}
\]

(5.15)

\[
\begin{bmatrix}
    s\theta_2 & 0 & c\theta_2 & 0 \\
    -c\theta_2 & 0 & s\theta_2 & 0 \\
    0 & -1 & 0 & 0 \\
    0 & 0 & 0 & 1
\end{bmatrix}
\]

(5.16)

\[
\begin{bmatrix}
    -s\theta_3 & 0 & c\theta_3 & 0 \\
    c\theta_3 & 0 & s\theta_3 & 0 \\
    0 & 1 & 0 & L_3 \\
    0 & 0 & 0 & 1
\end{bmatrix}
\]

(5.17)
TABLE 5.4: D-H PARAMETER OF THE HUMAN ARM

<table>
<thead>
<tr>
<th>JOINT i</th>
<th>a_i (mm)</th>
<th>( \alpha_i ) (deg)</th>
<th>d_i (mm)</th>
<th>( \theta_i ) (deg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>L_1(=0)</td>
<td>-90°</td>
<td>0</td>
<td>( \theta_1 )</td>
</tr>
<tr>
<td>2</td>
<td>0</td>
<td>-90°</td>
<td>0</td>
<td>( \theta_2 - 90° )</td>
</tr>
<tr>
<td>3</td>
<td>0</td>
<td>90°</td>
<td>L_3</td>
<td>( \theta_3 + 90° )</td>
</tr>
<tr>
<td>4</td>
<td>0</td>
<td>-90°</td>
<td>L_4(=0)</td>
<td>( \theta_4 )</td>
</tr>
<tr>
<td>5</td>
<td>0</td>
<td>90°</td>
<td>L_5</td>
<td>( \theta_5 )</td>
</tr>
<tr>
<td>6</td>
<td>L_6(=0)</td>
<td>-90°</td>
<td>0</td>
<td>( \theta_6 + 90° )</td>
</tr>
<tr>
<td>7</td>
<td>L_7</td>
<td>90°</td>
<td>0</td>
<td>( \theta_7 )</td>
</tr>
</tbody>
</table>

\[
3A_4 = \begin{bmatrix} c\theta_4 & 0 & -s\theta_4 & 0 \\ s\theta_4 & 0 & c\theta_4 & 0 \\ 0 & -1 & 0 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix}
\] (5.18)

\[
4A_5 = \begin{bmatrix} c\theta_5 & 0 & s\theta_5 & 0 \\ s\theta_5 & 0 & -c\theta_5 & 0 \\ 0 & 1 & 0 & L_5 \\ 0 & 0 & 0 & 1 \end{bmatrix}
\] (5.19)

\[
5A_6 = \begin{bmatrix} -s\theta_6 & 0 & -c\theta_6 & 0 \\ c\theta_6 & 0 & -s\theta_6 & 0 \\ 0 & -1 & 0 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix}
\] (5.20)

\[
6A_7 = \begin{bmatrix} c\theta_7 & 0 & s\theta_7 & L_7c\theta_7 \\ s\theta_7 & 0 & -c\theta_7 & L_7s\theta_7 \\ 0 & 1 & 0 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix}
\] (5.21)

As the transformation matrices are obtained, the trajectory of the hand for each kind of task is calculated using the Eq. (5.22).

\[
A_{\text{hand}}(\theta) = 0A_1^1A_2^2A_3^3A_4^4A_5^5A_6^6A_7
\] (5.22)

After identification of the amputee’s intended activity, the desired trajectory of the hand w.r.t. a fixed shoulder coordinate system are identified to realize the intended activity. The angular positions of 3 DOF shoulder and elbow motions used to identify
the intended activity are used to define the phase of the tasks. For each task, only one motion among the four is selected whose angular position is almost linearly varied during the initial to final state of the task. The angular position of this motion is used to determine the trajectory of the hand. In this study, shoulder flexion-extension angle is used for task (1, 9, and 10); elbow flexion-extension angle is used for task 2, and shoulder abduction-adduction angle is used for task (3-8) to calculate the hand trajectory.

5.5 Kinematics Based Controller (KBC)

For a given Shoulder \( A_{\text{Hand}} \), the trajectory of the hand w.r.t. a fixed shoulder coordinate system, inverse kinematics technique is used to find out the desired value of \( \theta_5 \), \( \theta_6 \), and \( \theta_7 \). Eq. (5.22) can be written as:

\[
^4A_5^5A_6^6A_7 = (^0A_1^1A_2^2A_3^3A_4)^{-1} \text{Shoulder } A_{\text{Hand}}
\]  

\[
^4A_5^5A_6^6 = \begin{bmatrix}
-c\theta_5s\theta_6c\theta_7 - s\theta_5s\theta_7 & -c\theta_6c\theta_7 & -c\theta_5s\theta_6s\theta_7 + s\theta_5c\theta_7 \\
-s\theta_5s\theta_6c\theta_7 + c\theta_5s\theta_7 & -s\theta_6c\theta_7 & -s\theta_5s\theta_6s\theta_7 - c\theta_5c\theta_7 \\
c\theta_6c\theta_7 & -s\theta_6 & c\theta_6s\theta_7 \\
0 & 0 & 0
\end{bmatrix}
\]  

Equating the components of both sides of Eq. (5.25),

\[
-c\theta_5s\theta_6c\theta_7 - s\theta_5s\theta_7 = p_{11}
\]  

\[
-s\theta_5s\theta_6c\theta_7 + c\theta_5s\theta_7 = p_{21}
\]
\begin{align*}
\cos \theta_6 \cos \theta_7 &= p_{31} & (5.28) \\
-\cos \theta_5 \cos \theta_6 &= p_{12} & (5.29) \\
-\sin \theta_5 \cos \theta_6 &= p_{22} & (5.30) \\
-\sin \theta_6 &= p_{32} & (5.31) \\
-\cos \theta_5 \sin \theta_6 \sin \theta_7 + \sin \theta_5 \cos \theta_7 &= p_{13} & (5.32) \\
-\sin \theta_5 \sin \theta_6 \sin \theta_7 - \cos \theta_5 \cos \theta_7 &= p_{23} & (5.33) \\
\cos \theta_5 \sin \theta_7 &= p_{33} & (5.34) \\
-\cos \theta_5 \sin \theta_6 \sin \theta_7 - \sin \theta_5 \cos \theta_7 &= p_{14} & (5.35) \\
-\sin \theta_5 \sin \theta_6 \sin \theta_7 + \cos \theta_5 \cos \theta_7 &= p_{24} & (5.36) \\
\cos \theta_6 \sin \theta_7 + L_5 &= p_{34} & (5.37) \\

\text{Using the Eq. (5.31), two possible values of } \theta_6 \text{ can be calculated as:}
\begin{align*}
\theta_6 &= \begin{cases} 
\arcsin(-p_{32}) \\
\left[-\pi + \arcsin(-p_{32})\right]
\end{cases} \quad (5.38)
\end{align*}

\text{Using the above equations the values for } \theta_5 \text{ and } \theta_7 \text{ can be calculated as:}
\begin{align*}
\begin{cases} 
\theta_5 &= \arctan \left(\frac{-p_{22}}{c \theta_6}, \frac{-p_{12}}{c \theta_6}\right) \\
\theta_7 &= \arctan \left(\frac{p_{33}}{c \theta_6}, \frac{p_{31}}{c \theta_6}\right)
\end{cases} \quad (5.39) \\
\begin{cases} 
\theta_5 &= \arctan(2(p_{13}, -p_{23})) \\
\theta_7 &= 0
\end{cases} \quad (5.40)
\end{align*}

\text{Since the range of motion is defined for the prosthetic arm } (-30^\circ \leq \theta_6 \leq 30^\circ), \text{ it is possible to find out only one desired value of } \theta_6. \text{ And the values of } \theta_5 \text{ and } \theta_7 \text{ are calculated using Eq. (5.41) and (5.42), respectively. As the desired angle is obtained and actual angle is measured from the actuator angular position sensor, a PD controller is used to control the corresponding prosthetic arm actuator.}
\begin{align*}
\theta_5 &= \arctan \left(\frac{-p_{22}}{c \theta_6}, \frac{-p_{12}}{c \theta_6}\right) \quad (5.41) \\
\theta_7 &= \arctan \left(\frac{p_{33}}{c \theta_6}, \frac{p_{31}}{c \theta_6}\right) \quad (5.42)
\end{align*}
5.6 EXPERIMENTS

Experiments were carried out to evaluate the effectiveness of the proposed controller for the AE prosthetic arm. Each actuator of the prosthetic arm is equipped with an incremental optical position encoder for angular position feedback. A personal computer with an interface card (RIF-171-1, Contec), an EMG signals amplifier (MEG 6108, Nihon Kohden) with an input box (JB-620J, Nihon Kohden), 5 motor drivers (iMDs03, iXs Research Corp.), and a DC power supply were used to carry out the experiments. A VICON motion capture system was also used in this study.

5.6.1 EBC FOR ELBOW AND HAND MOTION

In order to verify the performance of the prosthetic arm using the EMG signals based controller (EBC), experiments were carried out with a normal healthy human male subject (age 30 years). A schematic diagram of the experimental setup is shown in Fig.5.7. Four kinds of EMG signals were measured from the subject arm by using surface electrodes (NE-121J, Nihon Kohden). The raw EMG signals were amplified and sampled at a rate of 2 KHz. The RMS value of the EMG signals were then calculated (number of samples in a segment, \(N=100\)) and used as input information for the controller. The experimental results are shown in Figs. 5.8-5.10.

![Figure 5.7: Experimental Setup for EMG Signals based Controller.](image-url)
**Figure 5.8**: Experimental Results - Elbow Flexion & Extension Motion.
**FIGURE 5.9:** Experimental Results – Hand Grasp & Release Motion.

**FIGURE 5.10:** Experimental Results – Combination of Elbow & Hand Motion.
In order to verify the effectiveness of the controller with different EMG signals pattern, four experimental results are included under each experimental result caption. For the ease of demonstration only EMG signals of short head of biceps and long head of triceps are depicted in Figs. 5.8, which show that the activation of the biceps muscle results the elbow flexion motion, and extension motion is generated when triceps is activated. Figures 5.9 show that hand grasp motion is generated when both of the biceps and triceps muscles are activated simultaneously, and release motion is generated when both of them are not working. Moreover, it is also evident from Figs. 5.10 that both of the elbow and hand motion can be controlled using biceps and triceps muscles EMG signals.

5.6.2 Capturing Human Arm Motion – VICON Motion Capture System

![Figure 5.11](image)

**Figure 5.11**: Experimental Setup for VICON Motion Capture System.

Although the amputee’s shoulder and prosthesis elbow kinematics are supposed to be used as input signals for the proposed prosthetic arm controller, in this experiment, shoulder and elbow kinematics of the normal human are used to verify the controller technique. It is expected that the amputee’s can control the prosthesis elbow motion using EBC like natural human arm which is also shown in Figs. 5.8 to 5.10. In this study, the arm kinematics was acquired with a motion capture system. A commercially available VICON motion capture system (Vicon Inc.) was used to capture the shoulder flexion-extension, abduction-adduction, internal-external rotation,
elbow flexion-extension, forearm supination-pronation, wrist flexion-extension and radial-ulnar deviation angle of the healthy subjects while performing the selected tasks. The selected tasks have already listed in Table 5.3. A schematic diagram of the experimental setup for VICON motion capture system is shown in Fig. 5.11.

![Figure 5.11: Schematic Diagram of Experimental Setup](image)

**FIGURE 5.12:** Location of the Reflective Markers.

The system has ten cameras, each capable of recording at 240 Hz with images of 659x493 pixel resolution. A total of 22 reflected markers were placed on the body of the subject as shown in Fig. 5.12. The arm joint kinematics of the two subjects while performing the selected activities was collected at a sampling frequency of 60 Hz. Figures 5.13(a)-(b) show that a subject performing two different tasks during the experiment. During daily life activities, human arm performed various actions in either standing or sitting body posture depending on the nature of the activity. During the experiment, the subjects were instructed to perform the task in either standing or sitting position depending on the nature of the task. An unconstrained environment was maintained for the subject to pick up or move various small objects depending on the nature of the tasks. Subjects performed the activities using their right hand without any external load on their hand. During each task, the subject arm action was recorded by a video camera which was used during post processing of VICON data. For each task, the arm joint angles were calculated based on the Cartesian coordinates of each marker. This transformation is performed by VICON workstation using an inherent body model and used the anthropometric data of the subjects arm. Arm joint angles for two subjects during performing each task are depicted in Fig. 5.14 as example. It is evident from the experimental results as shown in Fig. 5.14; the arm joint angle patterns for each task can be varied from subject to subject. Moreover, it can be noted that the angle patterns of each task can be varied slightly during different attempts for the same subject.
FIGURE 5.13: Capturing Subject Arm Motion While Performing Different Tasks.
5.6.3 Task Classifier and KBC for 2 DOF Wrist and Forearm Motion

The effectiveness of the task classifier and task oriented kinematics based controller (KBC) were verified by using shoulder and elbow kinematics measured by the VICON motion capture system. The flowchart for the control of prosthesis forearm and wrist motion is shown in Fig. 5.15. The shoulder and elbow kinematics ($\theta$, $\dot{\theta}$ and $\ddot{\theta}$) is used as input for the task classifier.
FIGURE 5.15: Flowchart for the Control of Prosthesis Forearm and Wrist Motion.

The unit of the input variables for the artificial neural network (ANN) based task classifier is chosen as degree, rad/sec and rad/sec$^2$ for the angle, velocity and acceleration, respectively. The angular velocity and angular acceleration was calculated from the angle values obtained from VICON motion capture system. The ANN based task classifier was operated in two stages: training stage and evaluation stage. A part of kinematic data of the two subjects for all selected tasks was used during the training stage to adjust the weight values for the classifier. The adjusted weight values are used for the evaluation stage. The remaining kinematic data of two subjects for all selected task (which were not used during the training stage) were used for the evaluation stage to evaluate the accuracy of the classifier. Training stage begins with randomly selected initial weight values that exist between 0 and 1. Weight updating was performed for each sample of the kinematic data used for the training stage and stopped after 500 epochs. For the sample of input variables chosen from Task-N, the desired output of $O_N$ was selected to be 1. And desired output from the remaining output layer neurons was set for 0. Since the target outputs are selected, an evaluation function is defined which is shown in Eq. (5.43):

$$E = \frac{1}{2} \sum_{i=1}^{N^3} (y_d - y)^2$$  \hspace{1cm} (5.43)

where, $y_d$ and $y$ are the desired output and actual output from the output neuron, respectively. Then according to the error back-propagation learning algorithm, the weight values of $v_{jk}$ and $w_{ij}$ are updated by an amount $\Delta v_{jk}$ and $\Delta w_{ij}$, respectively, using the Eq. (5.44) and (5.45):

$$\Delta v_{jk} = -\xi \frac{\partial E}{\partial v_{jk}}$$  \hspace{1cm} (5.44)

$$\Delta w_{ij} = -\xi \frac{\partial E}{\partial w_{ij}}$$  \hspace{1cm} (5.45)
where, $\xi$ is the learning coefficient. After performing the training stage, final weight values are defined for the ANN based task classifier. To evaluate the performance of the task classifier as well as the KBC, experiments were carried out with the kinematic data which were not used during the training stage. The average success rate of the designed classifier is about 85%. After classification of the task, the desired trajectory of the hand for the classified task was measured using the angular position of the predefined shoulder or elbow motions (which define the state of the selected task). The KBC calculates the desired wrist and forearm angles and the actual angles are measured by the angular position sensor. Finally the necessary torque commands generated by the KBC send to the corresponding motors of the prosthetic arm via the motor drivers.

The experimental results for six kinds of tasks are shown in Figs. 5.16-5.21. For each kind of tasks two experimental results are depicted. The input signals, classifier accuracy, and the corresponding wrist angles for each of the experimental results are shown in (i), (ii), and (iii), respectively. For the ease of demonstration, only shoulder and elbow angles are depicted in (i). The desired wrist and forearm angles shown in (iii) are the angles calculated by the KBC. The proportional derivative controller of the KBC is used to generate the desired trajectory and the experimental results shows that the prosthetic arm can follow the desired trajectory perfectly. Figures 5.16 and 5.17 show, the prosthetic arm can generate the desired wrist and forearm angles. Figures 5.18 and 5.19 shows that the forearm pronation motion starts after reaching the hand at a certain shoulder and elbow position to fulfill the objective of the task. Furthermore, Fig. 5.21 shows that the proposed controller can generate similar hand movement for the repetitive movement of the shoulder and elbow as usually happened during shaving task. In the experimental results, as the task classifier identified different tasks due to the classifier inaccuracy, there is sometimes sudden change of wrist and forearm angles (i.e., error) for a short time. In case of the tasks-2, 3 and 4 (Figs. 5.16-5.18), the amount of error varies between 0°-6° and can be considered as a negligible amount practically. The amount of error varies between 0°-10° in case of the task-5 (Fig. 5.19) and is expected not to affect the task significantly at the point of classifier inaccuracy. The maximum amount of error occurred in case of tasks-7 and 8 (Figs. 5.20-5.21) and the amount of error is about 18°. However, increasing the classifier accuracy will inevitably reduce the amount of error.
**Figure 5.16**: Experimental Results – Eat with Spoon (Task-2).
FIGURE 5.17: Experimental Results – Brush Teeth (Task-3).
(i) Input – Sh. and Elb. Kinematics

(ii) Classifier Accuracy

(iii) Output – Wrist Angles

**FIGURE 5.18:** Experimental Results – Comb Hair (Task-4).
FIGURE 5.19: Experimental Results – Drink a Glass of Water (Task-5).
FIGURE 5.20: Experimental Results – Pick a Phone (Task-7).
FIGURE 5.21: Experimental Results – Shave (Task-8).
6.1 CONCLUSIONS

The design and control strategy of two kinds of robotic prosthesis is described in this thesis. The ultimate intention of the proposed implantable power assist robotic prosthesis is to assist daily life motion for elderly and physically disabled persons from inside of the body. Moreover, in order to increase the mobility and to improve the quality of life of above-elbow amputees in their daily life activities, a 5 DOF transhumeral prosthesis is also proposed in this thesis.

In chapter 2, a brief description about the human arm anatomy is provided. The motions provided by the human shoulder joint, elbow complex, wrist and hand, and the corresponding range of motions are described here. The information provided in this chapter is vital for the understanding of the following chapters.

In chapter 3, a new concept of implantable power assist prosthesis (i.e., inner skeleton robot) has been proposed and a prototype 2 DOF inner skeleton robot has been introduced to assist elbow flexion-extension and forearm supination-pronation motion from inside of the body for physically weak persons. The controllability of the designed joint actuator has been verified according to the motion intention of the user with applying a maximum of 3.0 ampere current that generates 0.1 Nm torque. The proposed system is supposed to be implanted inside of the human arm like the elbow arthroplasty and controlled by intramuscular EMG signals. Implantable Titanium coated flexible wire with needle tip can be expected to measure the intramuscular EMG signals for the proposed system. To keep the bone arrangement similar to the normal human elbow complex after the elbow arthroplasty and to transmit the assist
power to the radius and ulna bone, a T-mechanism has been introduced for the proposed prosthesis. As the ulna bone does not cross over the radius bone during the supination motion, it is not necessary to keep the length of the two ends of the T-link to be the same. The ratio of length of the ends of T-link can be changed according to the users bone geometry. The proposed inner skeleton robot can be expected to be an actuated artificial elbow joint for the future generation and supposed to act as a complete functional substitution of the normal human elbow complex.

In chapter 4, the mechanical design of the transhumeral prosthesis is described. Currently available transhumeral prostheses have failed to gain wide acceptance among the amputees society. One of the main factors that cause the loss of interest of the available prosthetic arm is their low functionality. It is difficult to generate natural human like arm motion using the available limited DOF prosthetic arm. In order to increase the mobility of AE amputees in their daily life activities, a 5 DOF transhumeral prosthesis using conventional DC motors is presented in this study. A modified T-mechanism is also introduced here for the prosthesis forearm motion. The stump arm holder (i.e., socket) is designed as a rectangular shape for the ease of experimental procedure. However, it can be designed as a half spherical structure like available expensive cosmetic prosthesis. The proposed design ensures that the range of motion of the prosthetic arm can be increase or decrease based on the user recommendation.

In chapter 5, the control strategy of the designed transhumeral prosthesis is proposed. The controller is designed based upon the assumption that the amputee’s stump arm consist the biceps and triceps muscles and those can be activated by amputee’s intension. As a result, the proposed controller become invalid for the user’s who doesn’t have any active biceps and triceps muscles in their stump arm. In this study, the EMG signals based controller (EBC) is designed and evaluated using the EMG signals of the normal human arm. Experimental results show the effectiveness of the proposed EBC. However, few uneven motions have also found during experiments due to stochastic nature of muscles EMG signals. Regarding to the VICON motion capture system, experiments were carried out to measure the human shoulder, elbow and wrist angles for the selected tasks. At the very beginning, the captured data were analyzed and found that the human forearm and wrist angle patterns varied excessively among the different subjects. As a result, during estimating the hand trajectory, the VICON captured wrist and forearm angles were
not considered. The wrist and forearm angles are estimated to generate the desired trajectory for the selected tasks. The data file obtained from VICON WORKSTATION consist arm joint angle values in degree. The unit of the shoulder and elbow joint angles is finally converted into rad/sec and their corresponding angular velocity and angular acceleration were calculated. A part of the joint kinematics data were then used to train the ANN based task classifier and the remaining data were used to evaluate the effectiveness of the task classifier as well as to evaluate the effectiveness of the KBC.

### 6.2 Future Works

The proposed power assist prosthesis is expected to be an active implantable device which requires power source to function from inside of the body. Although the power source for the implantable actuator is still undefined, Transcutaneous Energy Transfer (TET) system [110]-[111] or implantable battery with high power density can be considered as a power source for the proposed system. Moreover, from the standpoint of safety, the material for the proposed system and the insulation of the actuator stator and rotor parts lead the necessity of much research on it.

Regarding to the design of the transhumeral prosthesis, the intension to place the elbow motor in the proximal part of the forearm not only ensures sufficient space for stump arm to fit into the stump arm holder, but also it is expected to place part of the motor drivers and microcontroller into the holder. Aluminum alloy were used to fabricate the parts of the prosthesis. Using Magnesium alloy to fabricate the prosthesis parts will inevitably reduce the weight of the prosthesis as its specific density is about 65% than that of the Al alloy. The maximum amount of torque generated by the selected actuators for the prosthetic arm is listed in chapter 4. Considering the reduced amount of joint torque can also lead to a reduction of prosthesis weight. Moreover, light weight polymer material can be used in future to reduce the weight of the prosthesis. To make the appearance closer to a natural human arm, it is essential to cover the prosthetic arm with an artificial skin in near future. The type and installation procedure of the battery and microcontroller in the prosthetic arm were beyond the scope of this research work. However, lithium polymer batteries can be expected as a good candidate for the power source of the prosthetic arm due to their light weight.
and high energy density properties. Future works can also include placing the batteries on a waist belt and replacing the rectangular socket of the designed prosthesis with a half spherical structure similar to that employed in commercial prostheses.

Regarding to the controller part of the prosthetic arm, in this study, 10 daily life activities are considered to evaluate the concept of the proposed control technique and the classifier is trained with the 2 different subjects arm kinematics. However, much research can be carried out to include more daily life activities and to increase the success rate of the classifier. In the natural human arm, in case of the tasks like drinking water, forearm pronation motion is generated when the glass touches the lip. To imitate the natural human-like hand motion, in this study, forearm pronation motion is started after the hand reached at a certain level of shoulder and elbow position. In order to recognize the exact situation (glass touches the lip) to pronate the forearm, placing some sensors in the palm of the prosthetic hand is expected to bring out more impressive performance of the prosthesis. The VICON motion capture data is used in this study to evaluate the controller effectiveness. In order to measure the kinematics data of the user’s using wearable sensors; it is very essential to carrying on the research for the development of a 3D sensor that can measure the shoulder kinematics of the user’s arm.
DESIGN FUZZY MEMBERSHIP FUNCTION

A.1 THEORETICAL APPROACH TO DESIGN FUZZY MEMBERSHIP FUNCTION

In this research work, root mean square (RMS) value of user’s muscles EMG signals are used as the input information to control the proposed robotic prostheses. Three kinds of fuzzy linguistic variable (Zero: ZO, Positive Small: PS and Positive Big: PB) are prepared for each of the muscles described in this thesis. Two kinds of nonlinear functions: sigmoid \((f_s)\) and gaussian \((f_g)\) functions are used as the fuzzy membership function for each of the muscles. Let us consider the muscle \(n\) and its membership function as shown in Fig. A.1. The PB and ZO are expressed by the sigmoid function and the PS are expressed by the gaussian function. The calculation of the threshold value \((\omega_o)\) and weight \((\omega_i)\) is carried out as given below:

![Figure A.1: Membership Functions for Muscle-n.](image-url)
As shown in Fig. A.1,

$X_1 =$ Maximum RMS value of EMG signal of muscle-n

$X_0 =$ Minimum RMS value of EMG signal of muscle-n

The value of $X_2$ is calculated using Eq. (A.1)

$$X_2 = \frac{X_0 + X_1}{2} \quad (A.1)$$

$H\_RATE = 0.98$

$M\_RATE = 0.5$

$L\_RATE = 0.2$

For Positive Big ($f_b$):

$$w_i = (x_i, H\_RATE, x_2, L\_RATE) \quad (A.2)$$

$$w_i = \ln \left[ \frac{1}{H\_RATE} - 1.0 \right] / (x_2 - x_1) \quad (A.3)$$

$$w_0 = (x_i, H\_RATE, x_2, L\_RATE) \quad (A.4)$$

$$w_0 = (-\ln(\frac{1}{H\_RATE} - 1.0) - (w_i \times x_1)) \quad (A.5)$$

For Positive Small ($f_G$):

$$w_0 = -x_2 \quad (A.6)$$

$$w_i = (x_{1G}, M\_RATE, w_0) \quad (A.7)$$

$$x_{1G} = x_2 + \frac{(x_1 - x_2)}{2} \quad (A.8)$$

$$w_i = \sqrt{\frac{(w_0 + x_{1G})^2}{\ln(\frac{1}{M\_RATE})}} \quad (A.9)$$

For Zero ($f_z$):

$$w_i = (x_0, H\_RATE, x_2, L\_RATE) \quad (A.10)$$
\[
\begin{align*}
  w_i &= \ln \left[ \frac{1}{\frac{H\_RATE}{1} - 1} \right] \left/ \left( x_2 - x_o \right) \right. \\
  w_0 &= (x_o, H\_RATE, x_2, L\_RATE) \\
  w_0 &= (-\ln\left[ \frac{1}{H\_RATE} - 1.0 \right] - (w_i \times x_o))
\end{align*}
\] (A.11, A.12, A.13)

### A.2 MATLAB Program to Demonstrate Fuzzy Membership Function

The following program is developed to design fuzzy membership functions and simulate their shapes before apply to the controller parts for the proposed prostheses. This program is just a template which needs only modification of the RMS values to be used for different muscles.

```matlab
%Check_EMG_RMS_Fuzzification.m

emg = 0:0.001:1;

w_5 = 0.75;  
w_3 = 0.5;   
w_1 = 0.25;  
w_4 = w_5 - w_3; 
w_2 = w_3 - w_1;

h_rate = 0.98;  
m_rate = 0.4;   
l_rate = 0.02;

wi_h_00 = (log((1.0/h_rate - 1)/(1.0/l_rate - 1))) / (w_3 - w_5); 
wo_h_00 = ((-1.0)*log(1.0/h_rate - 1.0) - wi_h_00 * w_5);

wi_h_01 = sqrt((-w_3 + (w_3+w_4)/2)^2)/(log(1.0/m_rate));  
wo_h_01 = -w_3;

wi_h_02 = (log((1.0/h_rate - 1)/(1.0/l_rate - 1))) / (w_3 - w_1);  
wo_h_02 = ((-1.0)*log(1.0/h_rate - 1.0) - wi_h_02 * w_1);

us_h = wo_h_00 + wi_h_00 .* emg;  
ug = (wo_h_01 + emg) ./ wi_h_01;  
us_s = wo_h_02 + wi_h_02 .* emg;
```
```matlab
fs_h = (1.0 ./ (1.0 + exp(us_h * (-1.0))));
fg = exp((ug.*ug)*(-1.0));
fs_s = (1.0 ./ (1.0 + exp(us_s * (-1.0))));

figure;
plot(emg,fs_h,'b',emg,fg,'g',emg,fs_s,'r','LineWidth',1.5);
xLabel('Normalized RMS of EMG');
yLabel('Degree of Fitness');
title('Memebership Function of RMS EMG signal');
axis([0 1 0 1.1]);
ggrid on;
```
B.1 Amount of Wrist and Forearm Angles for Each Selected Tasks To Generate Desired Hand Trajectory

The daily life activities selected in this study for the proposed transhumeral prosthesis are listed in Table 5.3 and designated as Task1-Task10 in this thesis. A task oriented kinematics based controller is developed for the control of the proposed transhumeral prosthesis. In this research work, the desired hand trajectory to fulfill the objective of each task is estimated based on the nature of the task. In order to generate the required hand trajectory to accomplish the objective of the task, the desired wrist and forearm angles (w.r.t. the shoulder and elbow angles) are estimated for all kinds of selected tasks. The estimated wrist and forearm angles for each task are given here.

Here,
\[ \theta_1 = \text{Shoulder Flexion – Extension Angle} \]
\[ \theta_2 = \text{Shoulder Abduction – Adduction Angle} \]
\[ \theta_3 = \text{Shoulder Internal – External Rotation Angle} \]
\[ \theta_4 = \text{Elbow Flexion – Extension Angle} \]
\[ \theta_5 = \text{Forearm Supination – Pronation Angle} \]
\[ \theta_6 = \text{Wrist Radial – Ulnar Deviation Angle} \]
\[ \theta_7 = \text{Wrist Flexion – Extension Angle} \]
**Table B.1: Estimated Wrist and Forearm Angles for Task – 1**

<table>
<thead>
<tr>
<th>Task</th>
<th>Angle Define State of the Task</th>
<th>Angle</th>
<th>Initial Position</th>
<th>Final Position</th>
</tr>
</thead>
<tbody>
<tr>
<td>Task-1</td>
<td>$\theta_1$</td>
<td>$\theta_5$</td>
<td>-20°</td>
<td>0°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_6$</td>
<td>5°</td>
<td>0°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_7$</td>
<td>0°</td>
<td>-30°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_1$</td>
<td>0°</td>
<td>65°</td>
</tr>
</tbody>
</table>

Remarks: Wrist and forearm angles increase/decrease linearly with $\theta_1$

**Table B.2: Estimated Wrist and Forearm Angles for Task – 2**

<table>
<thead>
<tr>
<th>Task</th>
<th>Angle Define State of the Task</th>
<th>Angle</th>
<th>Initial Position</th>
<th>Final Position</th>
</tr>
</thead>
<tbody>
<tr>
<td>Task-2</td>
<td>$\theta_4$</td>
<td>$\theta_5$</td>
<td>20°</td>
<td>30°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_6$</td>
<td>-15°</td>
<td>5°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_7$</td>
<td>-30°</td>
<td>20°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_4$</td>
<td>30°</td>
<td>115°</td>
</tr>
</tbody>
</table>

Remarks: Wrist and forearm angles increase/decrease linearly with $\theta_4$

**Table B.3: Estimated Wrist and Forearm Angles for Task – 3**

<table>
<thead>
<tr>
<th>Task</th>
<th>Angle Define State of the Task</th>
<th>Angle</th>
<th>Initial Position</th>
<th>Final Position</th>
</tr>
</thead>
<tbody>
<tr>
<td>Task-3</td>
<td>$\theta_2$</td>
<td>$\theta_5$</td>
<td>-30°</td>
<td>10°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_6$</td>
<td>0°</td>
<td>-15°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_7$</td>
<td>-30°</td>
<td>-5°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_2$</td>
<td>10°</td>
<td>80°</td>
</tr>
</tbody>
</table>

Remarks: Wrist and forearm angles increase/decrease linearly with $\theta_2$
### Table B.4: Estimated Wrist and Forearm Angles for Task – 4

<table>
<thead>
<tr>
<th>Task</th>
<th>Angle Define State of the Task</th>
<th>Angle</th>
<th>Initial Position</th>
<th>Intermediate Position</th>
<th>Final Position</th>
</tr>
</thead>
<tbody>
<tr>
<td>Task – 4</td>
<td>$\theta_2$</td>
<td>$\theta_5$</td>
<td>-30°</td>
<td>0°</td>
<td>-30°</td>
</tr>
<tr>
<td>Task – 4</td>
<td>$\theta_2$</td>
<td>$\theta_6$</td>
<td>0°</td>
<td>-15°</td>
<td>15°</td>
</tr>
<tr>
<td>Task – 4</td>
<td>$\theta_2$</td>
<td>$\theta_7$</td>
<td>-30°</td>
<td>5°</td>
<td>25°</td>
</tr>
<tr>
<td>Task – 4</td>
<td>$\theta_2$</td>
<td>$\theta_2$</td>
<td>10°</td>
<td>100°</td>
<td>120°</td>
</tr>
</tbody>
</table>

Remarks: Wrist and forearm angles linearly increase/decrease with $\theta_2$ from initial to intermediate, and from intermediate to final position.

### Table B.5: Estimated Wrist and Forearm Angles for Task – 5

<table>
<thead>
<tr>
<th>Task</th>
<th>Angle Define State of the Task</th>
<th>Angle</th>
<th>Initial Position</th>
<th>Intermediate Position</th>
<th>Final Position</th>
</tr>
</thead>
<tbody>
<tr>
<td>Task – 5</td>
<td>$\theta_2$</td>
<td>$\theta_5$</td>
<td>0°</td>
<td>0°</td>
<td>-</td>
</tr>
<tr>
<td>Task – 5</td>
<td>$\theta_2$</td>
<td>$\theta_6$</td>
<td>-5°</td>
<td>0°</td>
<td>0°</td>
</tr>
<tr>
<td>Task – 5</td>
<td>$\theta_2$</td>
<td>$\theta_7$</td>
<td>-30°</td>
<td>-5°</td>
<td>-5°</td>
</tr>
<tr>
<td>Task – 5</td>
<td>$\theta_2$</td>
<td>$\theta_2$</td>
<td>10°</td>
<td>100°</td>
<td>-</td>
</tr>
</tbody>
</table>

Remarks: Wrist and forearm angles linearly increase/decrease with $\theta_2$ from initial to intermediate position. From intermediate position, for each degree of $+\theta_2$, pronation angle will increase 1°.

### Table B.6: Estimated Wrist and Forearm Angles for Task – 6

<table>
<thead>
<tr>
<th>Task</th>
<th>Angle Define State of the Task</th>
<th>Angle</th>
<th>Initial Position</th>
<th>Final Position</th>
</tr>
</thead>
<tbody>
<tr>
<td>Task-6</td>
<td>$\theta_2$</td>
<td>$\theta_5$</td>
<td>0°</td>
<td>-30°</td>
</tr>
<tr>
<td>Task-6</td>
<td>$\theta_2$</td>
<td>$\theta_6$</td>
<td>0°</td>
<td>0°</td>
</tr>
<tr>
<td>Task-6</td>
<td>$\theta_2$</td>
<td>$\theta_7$</td>
<td>-30°</td>
<td>-5°</td>
</tr>
<tr>
<td>Task-6</td>
<td>$\theta_2$</td>
<td>$\theta_2$</td>
<td>20°</td>
<td>100°</td>
</tr>
</tbody>
</table>

Remarks: Wrist and forearm angles increase/decrease linearly with $\theta_2$.
### TABLE B.7: ESTIMATED WRIST AND FOREARM ANGLES FOR TASK – 7

<table>
<thead>
<tr>
<th>Task</th>
<th>Angle Define State of the Task</th>
<th>Angle</th>
<th>Initial Position</th>
<th>Final Position</th>
</tr>
</thead>
<tbody>
<tr>
<td>Task-7</td>
<td>$\theta_2$</td>
<td>$\theta_5$</td>
<td>-30°</td>
<td>5°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_6$</td>
<td>0°</td>
<td>15°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_7$</td>
<td>-30°</td>
<td>0°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_2$</td>
<td>10°</td>
<td>80°</td>
</tr>
</tbody>
</table>

Remarks: Wrist and forearm angles increase/decrease linearly with $\theta_2$

### TABLE B.8: ESTIMATED WRIST AND FOREARM ANGLES FOR TASK – 8

<table>
<thead>
<tr>
<th>Task</th>
<th>Angle Define State of the Task</th>
<th>Angle</th>
<th>Initial Position</th>
<th>Final Position</th>
</tr>
</thead>
<tbody>
<tr>
<td>Task-8</td>
<td>$\theta_2$</td>
<td>$\theta_5$</td>
<td>-30°</td>
<td>-5°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_6$</td>
<td>0°</td>
<td>25°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_7$</td>
<td>-30°</td>
<td>5°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_2$</td>
<td>10°</td>
<td>65°</td>
</tr>
</tbody>
</table>

Remarks: Wrist and forearm angles increase/decrease linearly with $\theta_2$

### TABLE B.9: ESTIMATED WRIST AND FOREARM ANGLES FOR TASK – 9

<table>
<thead>
<tr>
<th>Task</th>
<th>Angle Define State of the Task</th>
<th>Angle</th>
<th>Initial Position</th>
<th>Final Position</th>
</tr>
</thead>
<tbody>
<tr>
<td>Task-9</td>
<td>$\theta_1$</td>
<td>$\theta_5$</td>
<td>30°</td>
<td>0°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_6$</td>
<td>0°</td>
<td>15°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_7$</td>
<td>-30°</td>
<td>-10°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\theta_2$</td>
<td>10°</td>
<td>50°</td>
</tr>
</tbody>
</table>

Remarks: Wrist and forearm angles increase/decrease linearly with $\theta_1$
**Table B.10: Estimated Wrist and Forearm Angles for Task – 10**

<table>
<thead>
<tr>
<th>Task</th>
<th>Angle Define State of the Task</th>
<th>Angle</th>
<th>Initial Position</th>
<th>Final Position</th>
</tr>
</thead>
<tbody>
<tr>
<td>Task-10</td>
<td>0&lt;sub&gt;1&lt;/sub&gt;</td>
<td>0&lt;sub&gt;5&lt;/sub&gt;</td>
<td>0°</td>
<td>35°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>0&lt;sub&gt;6&lt;/sub&gt;</td>
<td>0°</td>
<td>10°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>0&lt;sub&gt;7&lt;/sub&gt;</td>
<td>-30°</td>
<td>-10°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>0&lt;sub&gt;2&lt;/sub&gt;</td>
<td>15°</td>
<td>30°</td>
</tr>
</tbody>
</table>

Remarks: Wrist and forearm angles increase/decrease linearly with 0<sub>1</sub>
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