

**ÇUKUROVA UNIVERSITY
INSTITUTE OF NATURAL AND APPLIED SCIENCES**

MSc THESIS

Erkan ÖDEMİŞ

**DESIGN OF A CONTROL SYSTEM FOR REHABILITATION ROBOTIC
ARM ORTHESIS**

DEPARTMENT OF ELECTRICAL AND ELECTRONICS ENGINEERING

ADANA, 2015

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ABSTRACT

MSc THESIS

DESIGN OF A CONTROL SYSTEM FOR REHABILITATION ROBOTIC ARM ORTHESIS

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**ÇUKUROVA UNIVERSITY
INSTITUTE OF NATURAL AND APPLIED SCIENCES
DEPARTMENT OF ELECTRICAL AND ELECTRONICS ENGINEERING**

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Robotic upper extremity orthoses have been used for rehabilitation of patients with neuromuscular disorders since last two decades. The success of robotic arm orthoses in therapies is demonstrated by numerous clinical researches.

In this thesis, the design of a multi-mode model based controller and two DoF robotic arm orthosis in connection to a human arm mechanical model is presented. The designed system is implemented in simulation for compliant operation of robotic arm orthosis and human arm model. The controller is tested for multi-mode operation and its performance is evaluated for “Assistive” and “Resistive” rehabilitation modes. The tests yielded encouraging results for future works.

Key Words: Robotic Arm Orthosis, Model-Based Controller, Rehabilitation Robots.

ÖZ

YÜKSEK LİSANS TEZİ

REHABİLİTASYON AMAÇLI ROBOTİK KOL ORTEZİ İÇİN KONTROL SİSTEMİ TASARIMI

Erkan ÖDEMİŞ

**ÇUKUROVA ÜNİVERSİTESİ
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Robotik kol ortezleri, motor-kas becerilerini kaybetmiş hastaların tedavisinde son 20 yıldır kullanılan ve tedavideki başarıları klinik çalışmalarla kanıtlanmış cihazlardır.

Bu tez çalışmasında, çok-düzenli model tabanlı kontrolcü ile iki serbestlik derecesine sahip, insan kolu mekanik modeline uyarlanan bir robotik kol ortezi tasarımı yapılmıştır. Tasarım sonrası, robotik kol ortezinin ve bağlı olan insan kolu mekanik modelinin, benzetim ortamında uyumlu çalışması gerçekleştirilmiştir. Sistemin kontrolcüsü de “Destekleyici” ve “Zorlaştırıcı” rehabilitasyon düzenlerini içeren performans testlerinden geçmiştir. Tasarlanan ortez ve kontrolcü, gelecek çalışmalar için destekleyici sonuçlar vermiştir.

Anahtar Kelimeler: Robotik Kol Ortezi, Model-Tabanlı Kontrol, Rehabilitasyon Robotları.

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LIST OF SYMBOLS AND ABBREVIATIONS

q_d	:Joint Variable Desired Trajectory (Angular)
$\dot{q}_d$	: Joint Variable Desired Velocity (Angular)
$\ddot{q}_d$	: Joint Variable Desired Acceleration (Angular)
q	:Joint Variable Position (Angular)
$\dot{q}$	:Joint Variable Velocity (Angular)
$\ddot{q}$	:Joint Variable Acceleration (Angular)
τ	: Joint Torques
τ_{dis}	:Disturbance Torque
$\varepsilon(t)$	:Integral of Tracing Error
I_i	:Moment of Inertia of link i
$M(q)$	:Inertia Matrix
$V(q, \dot{q})$	: Coriolis/Centripetal Vector
$G(q)$	:Gravity Vector
K_p	:Proportional Gain
K_v	:Derivative Gain
K_i	:Integral Gain
K_p	:Proportional Gain
K_v	:Derivative Gain
K_i	:Integral Gain
θ	:Joint Variable Position (Angular)
$\dot{\theta}$	:Joint Variable Velocity (Angular)
$\ddot{\theta}$	: Joint Variable Acceleration (Angular)
$[\theta]^T$	:Transpose of θ
l_i	: Length of the i-th Link
l_{ci}	:Distance from the i-th Joint to the i-th Mass Center Position
a_i	: Length of the i-th Link
HAMM	:Human Arm Mechanical Model
DoF	:Degrees of Freedom

PM	: Pneumatic Muscle
GUI	: Graphical User Interface
CAD	: Computer Aid Design
AE	: Average Error
K	: Kinetic Energy
P	: Potential Energy
L	: Lagrangian
DM1	: Dynamic Model 1
DM2	: Dynamic Model 2

1. INTRODUCTION

Hundreds of people on earth lose their neuromuscular abilities due to accidents, illnesses or senility, and this situation affects their daily life negatively. Physiotherapy and rehabilitation are widely used methods for treating patients who lost their neuromuscular abilities.

Doing exercises with therapists is the traditional way in physiotherapy and rehabilitation treatment. But, there are some disadvantages such as not giving enough time for patients or a reduction in the quality of exercises due to busy working time of the therapist. Doing the exercises with robotic devices has started to be used as a new method to overcome the shortcomings of traditional treatment methods mentioned above. It has been proved with several researches that positive results have been obtained using robotic devices with rehabilitation treatments for patients who have lost their neuromuscular abilities (Volpe et al., 2001; Lum et al., 2002).

Using robots for rehabilitation has important advantages such as leaving enough time for patients, giving the opportunity of local treatment in suitable conditions, facilitating measuring and understanding the changes in patient's condition. Robotic rehabilitation devices are categorized according to their application place such as lower extremity, upper extremity or whole body.

Orthesis are assistant devices fixed to the body to increase the performance of the limbs which lost some part of their functions. Robotic arm orthesis for rehabilitation belong to the upper extremity class of the rehabilitation devices.

Patients are exercised for rehabilitation in assistive, resistive or repetitive modes in rehabilitation applications with robotic arm orthesis. The objective of this thesis is to design a control structure which is suitable for therapy modes and provides an opportunity for transition among these modes easily.

Mechanical design of robotic arm orthesis and the design of a controller in simulation has been performed in this thesis. Besides, the designed controller is aimed to apply assistive and resistive therapy modes.

The designed orthesis in this thesis has been tested attaching to a Human Arm Mechanical Model (HAMM). It has been aimed to see the effect of therapy modes

and mode transitions on human arm more realistically in this way. It has been considered that it would be more useful to make these modes and mode transitions in simulation setting since it would take longer and would be dangerous to test these therapy modes directly on humans.

In the second section of the thesis, studies on robotic arm orthosis are classified according to mechanical design and controller methods and analysed comparatively.

In the third section, mechanical design of HAMM and orthosis, CAD drawings and applied controller methods are described in detail.

In the fourth section, results and findings of the study are given and discussed.

In the last section, whole thesis is evaluated and planned improvements for the structure of the controller for the future are described.

2. PRELIMINARY WORKS

2.1. Orthosis Concept and Upper Extremity Orthosis

Orthosis are assistant devices installed to the body to increase the performance of the limbs which lost all or part of their functions. Robotic arm orthosis installed to the outside of a limb are also devices designed to make a limb move or do some exercises for rehabilitation by using some actuators. Robotic rehabilitation devices are categorized according to their place of use such as lower extremity, upper extremity or whole body orthosis (Gopura et al., 2009a). Robotic arm orthosis for rehabilitation belong to the upper extremity class of the rehabilitation devices.

The use of robotic orthosis for treatment, especially for the upper limbs began after 1980s and in the beginning of 1990 (Erlandson, 1995; Reinkensmeyer et al., 2000). Nowadays many studies have been made in this field with the advancements in the technology (Gopura et al., 2009b; Ball et al., 2007; Wolbrecht et al., 2009).

2.2. Review of Upper Extremity Robotic Arm Orthosis

Resrarchs on robotic arm orthoses can be reviewed under two main topics that is orthosis mechanical design and control method.

2.2.1. Mechanical Design

Robotic arm orthoses for rehabilitation can be studied under some fields according to mechanical designs, such as number of joints they have, actuators used for motion and power transmission methods.

A human arm has three Degrees of Freedom (DoF) at the shoulder (shoulder flexion/extension, shoulder abduction/adduction and shoulder internal/external rotation), two DoF at the elbow (elbow flexion/extension and elbow supination/pronation) and two DoF at the wrist (wrist flexion/extension and

radial/ulnar deviation) and totally has 7 degrees of freedom from shoulder to wrist as shown in Figure 2.1 (Gopura et al., 2009a). When orthosis are studied according to number of DoF they have, there are some orthoses such as SUEFUL-7 (Gopura et al., 2009b), ARMIN (Nef et al., 2007), INTELLIARM (Park et al., 2008), MEDARM (Ball et al., 2007) or PNEU-WREX (Wolbrecht et al., 2009) which have all degrees of freedom as human arm and there are also some orthoses such as WATOS (Rocon et al., 2007), W-EXOS (Gopura and Kiguchi, 2008) or ASSIST (Sasaki et al., 2005) which are designed for only wrist rehabilitation or wrist and elbow rehabilitation.



Figure 2.1. Human Arm Degrees of Freedom (Gopura et al., 2009a)

2.2.1.1. Actuators

Two different actuators such as Pneumatic Muscles (PM) and motors are used in robotic arm orthosis (Lo et al., 2012).

Pneumatic muscles are actuators which use air pressure for motion. One of the important features of the pneumatic muscles are the high power/weight ratio. With this advantage of PMs the weight of robots and the power need to move the robot can be reduced. Another advantage of pneumatic muscles is their flexible structure. They are the most similar actuators to the human muscle and they are also the most suitable actuators to move compliant with the patient thanks to this feature of the pneumatic muscles. There are disadvantages of pneumatic muscles such as non-linear movement characteristics and working at a low rate of accuracy however. There are many orthoses in literature designed using pneumatic muscle or pneumatic cylinder (Tsuji et al., 2015).

Tsagarakis and Caldwell (2003) have designed 7 DoF upper limb exoskeleton with PMs. Their primary aim in this study is designing an exoskeleton that has low mass, high power-weight ratio, high safety, natural compliance and low cost.

For providing low mass, they have primarily used aluminum and composite materials in the structure. The most important part of their study is actuators. They manufactured the PMs (Pneumatic Muscles) which were used in that study, lead off their previous studies (Davis et al., 2002). Using in-house manufacture instead of commercial version with similar characteristics which is produced by Festo, PMs permit greater control over dimensions, forces and general performance of the drivers. Produced torque and force by PMs are transmitted to the joints by cables and pulleys. For providing compliance control, they used an antagonistic pair of PMs. Also high tolerance of the actuator against mechanical misalignments makes the system easy for construction and keeps the complexity and cost low. PMs' flexibility and soft nature ensures the safety requirements for the user.

D. Sasaki et al. (2005) have developed the Active Support Splint driven by pneumatic soft actuator (ASSIST) to assist only wrist flexion motion of elderly or people in need of care. In that study, two (2) types of ASSIST are designed. First type is designed to increase the movable range and second type is designed to increase muscular endurance. The difference between two types is McKibben types of pneumatic rubber muscles. The McKibben type pneumatic rubber muscles are attached to type 1 to relieve the restraining feeling by releasing the palm appliance when the device is not operated.

Designers of the ASSIST used soft materials like rubber in the construction of the actuator, so the device has flexibility due to power source and the structure of the actuator. In each type of ASSIST, there is a rotary type of soft actuator in both sides of the device. But rotary type of soft actuators which are used in type 1 and type 2 have different structures. Type 1 also has a McKibben type of pneumatic rubber muscle in both sides. Structures and properties of both rotary types of soft actuators are described in the study.

Experimental studies with type 1 of ASSIST show that it can move the human wrist joint when the patient doesn't generate muscular force. Also, other

experiments present that type 2 of ASSIST prevents the muscular fatigue and increases the muscular endurance. In this research, there is also a study about human intention. With this study, ASSIST can be switched ON or OFF based on the measured flexion angle and an average threshold value (5°).

Klein et al. (2008) have designed another device for upper limb rehabilitation named BONES (Biomimetic Orthosis for Neurorehabilitation of Elbow and Shoulder). The importance of this study is the novel upper extremity rehabilitation exoskeleton design. BONES has 4 DoF (3 DoF at shoulder and 1 DoF at elbow) and uses pneumatic cylinders for creating motion. Biomimetic part of this study from supination/pronation motion at human wrist joint. At this joint, ulna and radius bones generate supination/pronation motion at wrist. Designers of BONES were inspired about motion generation by this method. Two rods that are activated by pneumatic cylinders generate motion at shoulder joint, like the motion of human wrist joint. Designers call this structure that generates motion in shoulder “Diamond Structure”. BONES also has 3 elastic elements which are parallel to the cylinders 1, 2 and 3 for providing weight support and gravity compensation.

BONES uses a new design, but also has a big and heavy structure. So it is not proper for home rehabilitation and it is not portable.

In another study, Wolbrecht et al. (2006) have designed PNEU-WREX (Pneumatic-Wilmington Robotic Exoskeleton) that is robotic version of WREX (Wilmington Robotic Exoskeleton) (Rahman et al., 2007).

WREX is a passive exoskeleton and has gravity compensation with elastic bands, but PNEU-WREX has 4 active DoF (clavicle rotation, shoulder flexion/extension, shoulder abduction/adduction and elbow flexion/extension) and it is actuated by Pneumatic cylinders. PNEU-WREX also uses four bar linkage mechanism for elbow joint, because different methods such as pulleys or gears are cumbersome and costly.

Balasubramanian et al. (2008) have designed Robotic Upper Extremity Repetitive Trainer (RUPERT). It is designed for assisting in repetitive therapy tasks related to daily life activities and driven by McKibben type pneumatic muscle actuators. RUPERT has 5 degrees of freedom. (shoulder flexion/extension, shoulder

abduction/adduction, elbow flexion/extension, forearm supination and elbow flexion). They used one pneumatic muscle for each joint, so, unlike the antagonistic muscle pairs, Rupert has one direction of motion at each joint.

For finding and designing the proper size length and attachment points for PMs, they used SIMM (Software for Interactive Musculoskeletal Modelling) simulation program. With this program, they developed a kinematic model of upper extremity and PM's features were designed based on this model. RUPERT doesn't have any gravity compensation. So the user of RUPERT performs the rehabilitation exercises in a natural setting. RUPERT had been tested over ten subjects, and some of the subjects had very good improvements.

Motors are other types of common actuators used in robotic arm orthoses. There are important advantages of motors such as being compatible with high-speed, sensitivity, advanced motion control. But these motors have some disadvantages like being heavy and bulky, needing maintenance and their structure is not suitable for moving compliant with patients since they produce high amount of torque. There are many studies which used motors as actuators in the literature (Gopura et al., 2009b; Perry et al., 2007).

Almost all studies in the literature use only electric motors or pneumatic muscles for actuation. However, a few studies use both electric motors and pneumatic muscles together in actuation system of the exoskeleton. One of the rare studies is IKerlan's Orthosis (IKO).

IKO has been developed by Martinez et al. (2008). IKO was designed for increasing human power during the activities of daily life. IKO has 5 active and 4 passive DoF. Active DoF was used to increase the workspace of the exoskeleton. IKO has 3 active DoF at shoulder (abduction, flexion and rotation) and 2 active DoF at elbow (flexion and forearm rotation). There is no DoF at wrist because designers of the IKO assumed that user could use his/her hand for grasping the objects.

At the beginning of the paper, authors indicated that one of their initial requirements is, if it is possible, to use non-conventional actuators. Therefore, forearm rotation joint was actuated by two pneumatic muscles. Conventional electric motors were used at other four joints.

For experimental works, a prototype of IKO which has 2 DoF at wrist joint was constructed. Elbow flexion joint was driven by electric motors, and forearm rotation joint was driven by pneumatic muscles. With experimental studies, some data were obtained about characteristics of electric motors and pneumatic muscles.

2.2.1.2. Power Transmission

Another important point in orthosis design is the method used for power transmission. Main methods in power transmission in orthosis design are gear systems and cables (Gopura et al., 2009a).

In designs which motors are placed on joints, power transmission is done with gear systems. This method results simple designs, but the moving parts of the orthosis become heavier. More torque and power are needed to move joints as a result.

The motors can be placed to fixed parts if cables are used for power transmission; this reduces the need of torque and power for motion by decreasing weight of moving parts. But designing an orthosis by using cables for power transmission is more complex than designing with gear systems.

Mao et al. (2011) have designed Cable-Driven Arm Exoskeleton (CAREX) for neural rehabilitation. CAREX has 5 DoF; three at shoulder (abduction, extension, internal-external rotation of upper arm), one at elbow flexion and extension, and one at forearm supination and pronation.

Most important part of the CAREX's design is that there is no rigid links or joints in the structure. Power transmission is supplied by cables. With the placement of motors away from joints and not using rigid links, mass and inertia of the exoskeleton is reduced.

3 cuffs are used in the CAREX. One of these cuffs is placed at shoulder, another one is placed at upper arm, and the last one is placed at the forearm. There are 7 cables in the CAREX structure to drive the exoskeleton. Four of these seven cables start from shoulder cuff, and terminate at upper arm cuff. These cables drive the 3 DoF at the shoulder. The remaining three cables terminate at the forearm cuff,

and these cables drive both the elbow flexion motion and forearm supination and pronation motion.

CAREX's lightweight structure is a major advantage. Also its design avoids the joint misalignment problem. But, total structure of the CAREX is too big, and it is not proper for home-rehabilitation.

Controller structure consists of two levels. High level force field controller plans the cable tensions to create a tunnel-like force field which is used to help the hand follow a nominal path around the hand. Lower level controller takes the commands from the higher level, and implements them to compensate the friction and other losses.

Experiments of CAREX have been performed with dummy robotic arm and 5 healthy subjects, and results showed that the controller can keep the hand closer to the nominal path.

In another study, Perry et al. (2007) have designed Cable-Actuated Dexterous Exoskeleton for Neurorehabilitation (CADEN-7).

CADEN-7 has 7 DoF (shoulder abduction–adduction, shoulder flexion–extension, shoulder internal–external rotation, elbow, forearm pronation–supination, wrist flexion–extension and wrist radial–ulnar deviation). The study which uses motors to drive the joints, power transmission has been achieved by cables because of its frictionless and without backlash nature.

For reducing the weight of moving parts, the first four motors which are heavier than remaining three, have been placed on the stationary base. With this method, weight of moving parts is reduced by %60.

CADEN-7 has a bulky structure. But it achieves %99 of the range of motion required to perform daily activities. Also this design presents a solution to singularity problems. These reasons make it one of the important studies in the literature.

2.2.1.3. Shoulder Joint Abduction Motion Problem

One of the most challenging subject about exoskeleton design is shoulder joint. The rotation center of the joint makes both rotation and translation during abduction movement (Gopura et al., 2009a). There are many studies on this situation.

One of those studies was made by Papadopoulos and Patsiaris (2007). The purpose of this study was to produce a solution to the problem in the shoulder joint stated above. First, the state of the shoulder joint was determined by coordinate measuring machine. As a result, it was found that rotation center of the shoulder joint moved 3 mm from starting point. Four bar link and Genova mechanism were added to the shoulder desing to solve this problem, so the rotation center of the shoulder joint achieved to make translation and rotation movements together.

. The success of the study was not tested since it was just at the designing process.

Kiguchi et al. (2003) have designed an exoskeleton with two degrees of freedom for shoulder rehabilitation. They have studied about shoulder rehabilitation since shoulder joint is very important for the activities of daily life. The exoskeleton has two degrees of freedom named abduction-adduction and flexion-extension. The most important purpose of this study was to make glenohumeral joint taking the change of turning center into consideration during the flexion movement. The arm holder in exoskeleton is connected to a linkwork mechanism. That mechanism prevents possible ill effects which are a result of joint incompatibility by regulating the distance between arm holder and turning center.

Electromyography (EMG) signals from 8 different points on the client makes the input data of controller. The purpose of selecting EMG signals as input data is to estimate the intention of client's movement/motion. Fuzzy neuro controller is used to achieve the adaptation to other clients.

Experiments with two healthy clients were carried out with and without exoskeleton. It was seen that EMG level drops when exoskeleton is used. It shows robot supports the client.

In another study about shoulder complex, Ball et al. (2007) have designed the MEDARM (Motorized Exoskeleton Device for Advanced Rehabilitation of Motor function). MEDARM has 6 DoF. The important specialty of MEDARM is to have 2 DoF at sternoclavicular joint (elevation/depression and protraction/retraction). MEDARM also has 3 DoF at shoulder joint (glenohumeral joint), and one DoF at elbow joint. For actuation, MEDARM uses motors, and it uses cables for power transmission and belt transmission system for reducing the inertia and maximizing the the power-to-weight ratio of the robot.

Despite the fact that the MEDARM enables to control whole shoulder complex, it has a huge and heavy structure. Also, it is only realized in the simulation environment.

Carignan et al. (2009) have designed the Maryland-Georgetown-Army (MGA) arm for shoulder and arm rehabilitation. In MGA arm, there is one DoF at scapula for shoulder elevation and depression motion. Also there are 3 DoF at shoulder and one DoF at elbow, so totally it has 5 DoF. Additionally, there is a passive joint for forearm supination/pronation motion. MGA arm uses DC motors to actuate the joints.

In this study, a composite structure controller is used and sets of joints are controlled separately according to the limbs to exercise. For example the shoulder joints are controlled together, elbow and scapula joints are controlled independently during the shoulder exercises. All subsystems of the controller can control the arm in admittance, impedance or position control mode depending on the dominance of the position or torque information.

2.2.1.4. Human Wrist Joint

Another challenging point in exoskeleton design is the wrist joint. There are two degrees of freedom in human wrist joint; radial ulnar deviation and flexion extension motion. Studies in the literature showed that two movements/motions in the human wrist joint occurs in two different axes which makes four different axes (Gopura et al., 2009a).

This situation is neglected in some studies (Perry et al., 2007). Some studies like SUEFUL-7 (Gopura et al., 2009b) took it into consideration. Radial/ulnar deviation and flexion/extension movements/motions are gathered at two different axes and an offset of 5 mm is left between two axes in SUEFUL-7.

2.2.2. Control Methods

Exoskeleton controllers could be classified in many ways. In this review, controllers are classified according to model based control systems, hierarchy based control systems and physical parameters based control systems (Anam et al., 2012). Also, some advance properties of the controllers are explained briefly.

2.2.2.1. Model Based Control Systems

Model based control strategies could be grouped into dynamic model based controllers and muscle model based controllers (Anam et al., 2012).

2.2.2.1.(1). Dynamic Model Based Controllers

Dynamic model is developed by designing fixed links and joints which are connecting these links like human body. This model tries to estimate the torque produced by inertia, gravitational, coriolis and centrifugal effects. Dynamic model can be developed by using mathematical model, system identification method and artificial intelligent methods.

Mathematical models are developed by modelling the system theoretically according to physical characteristics. Berkeley Lower Extremity Exoskeleton is one of the important studies in which dynamic model is used (BLEEX) (Kazerooni et al., 2005). The BLEEX only relied on its dynamic model to aid the user's movement, without any force or torque sensors to detect the interaction between the user and the exoskeleton. BLEEX is a device designed for rehabilitation for lower extremite. It is

used in this study since it has a different kind of controller system as mentioned above.

Another research in which dynamic model is used is the exoskeleton designed by Lee et al. (2014). This exoskeleton has two degrees of freedom at shoulder and one degree of freedom at the elbow which makes three degrees of freedom at all. The intention of motion from the user is sensed with the power sensors at the end-effector, this data makes the input of the controller system. The control of the exoskeleton in this study is done directly with the mathematical model.

System identification method is used when it is difficult to get a good dynamic structure with mathematical method. It is tried to estimate the dynamic parameters of the system according to specific input and output values.

Artificial intelligence which is the third method is chosen because of its success in solving non-linear problems. A neural network is trained in this research to estimate the torque by using the data from joint angle, joint speed and joint acceleration. The trained network is used as the inverse dynamic model of the system.

Another research which uses artificial intelligence method is done by Boukens and Boukabou (2015). The controller in this study consists of two parts such as robust part and neural network based control part. An artificial neural network is trained and used as a dynamic model of the system. The developed controller is tested on a structure with three joints successfully.

2.2.2.1.(2). Muscle Model Based Controllers

Muscle models in contrast to the dynamic models estimate the muscle moment as a function of muscle activation level and joint kinematics (Anam et al., 2012). In muscle model, the input is EMG signal and the output is the strength estimation.

There are two basic structures to construct a muscle model. One of them is parametric model, and the other one is non-parametric model. Parametric model is generally constructed using Hill-based muscle model.

Hill-based muscle model consists of contractile element, series element and parallel elements (Rosen et al., 2001). Hill model produces an output which is a function of EMG neurotic activity and muscle length. The formulation of Hill muscle model is complex and difficult, but it has a universal structure and it is not affected from to subject.

One of the studies about Hill muscle is the study made by Rosen et al. (2001). The exoskeleton consists of two joints, the purpose of which is to get the patient's movement/motion intention and assist client's motion. One of the joints is on the shoulder and the other one is on the elbow. Since the joint on the shoulder is fixed, the structure's only degree of freedom is the elbow joint.

Hill muscle model is used in the controller structure of the exoskeleton. Hill muscle model estimates the signal produced by the muscle using length and speed of the muscle and EMG signals and it produces a command signal. The input signals of the Hill muscle are EMG signals from primary flexor and extensor muscles, and external load and joint kinematic values from sensors.

Robot is operated in passive and active modes in the experimental studies. It is found that EMG signal level from the subject in active mode is lower than passive mode. That shows the robot provides motion assist.

Non-parametric model maps the input and output sets using neural network without using any muscle function (Gunasekara et al., 2012; Anam et al., 2012). Neural network is educated using known input and output values, so neural network can estimate an output according to a random input value.

Compared with each other, neural network model can make better estimations than Hill based model. But the success of the neural model depends on the tasks, when tasks, that are input sets, changes, the success rate of the neural network reduces.

One of the important examples of non-parametric muscle models is SUEFUL-7 which was designed by Gopura et al. (2009b). SUEFUL-7 is a robotic exoskeleton which has 7 degrees of freedom and is designed to assist the motion of upper limb extremity. EMG-based fuzzy-neuro control and muscle-oriented control method is used in the controlling structure of Sueful 7. Fuzzy-neuro control method

gave good results for robotic control in the previous studies. But muscle model control method is used with EMG-based fuzzy-neuro control since the control of the robot gets worse as the degree of freedom increases.

The primary input in the structure of SUEFUL-7 is EMG signal. EMG signals from 16 different points on the patient are obtained and these are used to estimate the motion intention of the patient. Forearm force, torque and hand force vectors of the patient are measured with EMG signals by sensors.

The control of SUEFUL-7 is established with EMG signals if the level of EMG signals from the client is high. The control is established with the information obtained from sensors if the level of EMG signals from the patient is low. On the other hand, the control is established with the combination of EMG signals and sensor information if the muscle activity of the client is middle-level.

A decrease in the level of EMG signals of subjects using exoskeleton in the experimental studies with SUEFUL-7. This shows that SUEFUL-7 assists the motion.

2.2.2.2. Hierachy Based Control Systems

One of the methods used in the classification of controllers is the hierarchy based control. Hierarchy based control can be studied in three groups such as task level, high level and low level (Anam et al., 2012). Task level control is the highest level controller which produces command signals according to tasks designed in the structure of the controller. The next controller structure is high level controller.

High level controller is responsible for the control of the force and position of the human-machine interaction according to the command signals from task level controller. High level controller is generally impedance and admittance controllers (Hogan et al., 2005).

Impedance controller is extended version of position controller. General equation of impedance control law is given in Equation 2.1 where $Z_{(s)}$ is impedance, $F_{(s)}$ is force and $v_{(s)}$ is velocity.

$$Z_{(s)} = \frac{F_{(s)}}{v_{(s)}} \quad (2.1)$$

It takes the velocity information as the input information, and it produces force as an output information. Force value which is an output information makes the input information, a reference value, for force controller which is a low level controller.

Force controller tries to make the force applied by exoskeleton similar or the same with the reference force.

Admittance controller which is another high level controller takes the force as an input, and produces position information as an output. General equation of admittance control law is given in Equation 2.2 where $A_{(s)}$ is admittance, $F_{(s)}$ is force and $v_{(s)}$ is velocity.

$$A_{(s)} = \frac{v_{(s)}}{F_{(s)}} \quad (2.2)$$

This position information is an input information for the position controller which is another low level controller. Position controller controls the joint angles of the exoskeleton according to the position information from admittance controller.

2.2.2.3. Physical Parameter Based Control Systems

Physical parameter based controllers can be classified under three groups such as position controller, torque/force controller and force interaction controller (Anam et al., 2012).

Position controller is to bring the angles of orthosis joints to the desired level. Position controllers are used as low-level controllers as mentioned before. PD position controller in MGA arm orthosis is used as low-level controller (Ball et al., 2007). PID position controller is used as low-level controller in RUPERT (Balasubramanian et al., 2008).

Torque/force controller which is another physical parameter based controller like position controller is also used as a low-level controller. The aim of the torque/force controller is to make the human exoskeleton interaction force zero or almost zero. Torque/force controller is used as a low-level controller in PNEU-WREX (Wolbrecht et al., 2006).

Force interaction controller which is another physical parameter based controller is used as a high-level controller in contrast to position controller and force controller. The function of the force interaction controller is to provide the proper help for patient to do functions of the controller, and to make exoskeleton interaction force zero in this process. Interaction forces are controlled with impedance and admittance controllers.

SUEFUL-7 and PNEU-WREX are orthoses which use impedance controllers as high level controller (Gopura et al., 2009b; Wolbrecht et al., 2006). Admittance controller is used as high-level controller in MGA arm (Ball et al., 2007).

2.2.2.4. Adaptive Control

One of the required specifications is the function of adaptability for versatile client use.

EMG signals may differ from client to client according to position of the electrodes. It may even differ on the same client according to tiredness and some other factors. Because of that, it is difficult to make an adaptive controller using EMG signals. Gopura et al. (2009b) made an adaptive study using neural network structure with EMG signals. In RUPERT, an iterative learning algorithm was used to have an adaptive controller algorithm (Balasubramanian et al., 2008).

If EMG signals, and some other factors such as mechanical structure of the muscle, and range of motion of the weak muscles are added to the controller system, a structure of control system which can be adapted exactly to patient will be produced.

2.3. Asistive and Resistive Rehabilitation Modes and Compliance Motion

Orthoses with active assistive function are devices which help patients with daily activities like reaching and grasping (Marchal-Crespo et al., 2009). Assistive exercises make patients complete the intended movements by applying external force. The physical assistance help patients complete the movement earlier, and the patients could make more effective practices this way.

One of the important things in assistive control strategies in exoskeleton control is to apply the assist to the patient as much as needed. Wolbrecht et al. (2008) made a study about it. The purpose of the study was to design a compliant controller which assists the client at the lowest level. The structure of the controller was tested on the PNEU-WREX (Wolbrecht et al., 2006).

PNEU-WREX is an exoskeleton which has 4 degrees of freedom, makes the actions with pneumatic actuators and uses coils to support the weight. PNEU-WREX is chosen since it is a device which could produce a lot of output force and had a wide range of motion.

In PNEU-WREX researchers call the structure of the controller which gives the minimum assist to the patient “assistance as needed”. In their hypothesis, the system which assists the patient to complete the intended movement/motion at a maximum level with the least assistance makes the patient benefit from the therapy at a high level.

The structure of general controller is model based adaptive control. The movement capability of the clients is transmitted to the model in real time using radial basis function representation.

Assistance-as-needed function of the controller structure is achieved by adding force reducing term to the adaptive controller structure which will reduce the power produced by the robot if the errors are small.

The success of the controller is tested by two different groups. The purpose of the first experiment, spatial tracking, was to prove that the device help patient finish the movement/motion. The purpose of the second experiment, side-to-side

experiment, was to test the robot's function of assistance-as-needed. Both experiments with the device had positive results.

Resistive mode in contrast to assistive mode aims to make patient spend more effort by breaking or making the movement more difficult (Marchal-Crespo et al., 2009). Orthosis makes the patient spend more effort by applying some resistive force directly against his movement while performing the action. Another method to achieve resistive mode is to make the movement by increasing errors. It is generally aimed to reduce the errors in assistive strategies. In this method, it is contrarily aimed to make the patient spend more effort to fix the errors by increasing errors.

One of the features that should be in rehabilitation orthoses is the high mechanical compliance (Morasso et al., 2009). That kind of orthoses has a feature which detects the intention of patient's movement and acts compliantly with the patient.

One of the most important things about mechanical compliance is that orthosis should have low-stiffness characteristics. A stiff-controlled orthoses make the rehabilitation exercises without patient's contribution. This reduces the medical effect of exercises greatly. Controller must be suitable, and orthoses should have low-friction and low-inertia features to be low-stiffness. Besides, low-stiffness orthosis are more reliable for patient's health.

2.4. Importance of Rehabilitation Modes and Simulation Studies

The function of upper-limb orthoses is to help patients get well soon and increase their muscle capabilities. Studies in the literature show that orthoses which provide therapy exercises at assistive/resistive mode therapy get successful results about the treatment of patients (Marchal-Crespo et al., 2009). For this reason, it is very important to add assistive or resistive mode to the controller for the success of the treatment. It is dangerous to test assistive/resistive modes and mode translations on patients directly and it may harm patients. So it is a safer method to test the performance of orthosis and controllers in a simulation environment, it is especially safer to test it on a human arm mechanical model (HAMM) first to see the effects of

them on human arm. Besides this method is more suitable since it lets us make changes on controller according to results. After succesful results obtained in simulation, it is possible to transfer the design into hardware applications.

3. MATERIALS AND METHOD

In this chapter, design stages of a 2 DoF robotic arm orthosis in connection to a Human Arm Mechanical Model (HAMM) and as well as a model based multi-mode workspace controller is explained. In addition, the simulation is also explained. First, mechanical design of orthosis and HAMM is explained in detail in this chapter. Then, how the designed mechanical models were transferred to Matlab is explained. The model which were created in Matlab and the blocks which create this model are shown in detail. Finally, the controller created in Matlab and Graphical User Interface (GUI) was explained in depth.

3.1. Orthosis Design

The design of the orthosis consists of two parts named controller design ve mechanical design. The mechanical design of orthosis and HAMM are made using Solidworks. Solidworks is a 3 Dimension Computer Aided Design (3D CAD) software¹ (Figure 3.1).

3.1.1. Mechanical Models

Human arm has 7 degrees of freedom from shoulder to wrist as mentioned in Chapter 2, in order to simplify the controller design, the orthosis in this thesis has 2 degrees of freedom. Dynamic model of an orthosis with two degrees of freedom is easier to get, and this is the reason why the design of the orthosis is done like that.

One of the things that should be taken into consideration is to prevent patient's joint misalignments with orthosis. So, HAMM on which orthosis will be tested is designed first, then appropriate orthosis for HAMM is designed. Motors are preferred as actuators for HAMM and orthosis since they are appropriate for advanced motion control. Gear systems are used to get a simpler mechanical design.

¹ <http://www.solidworks.com>

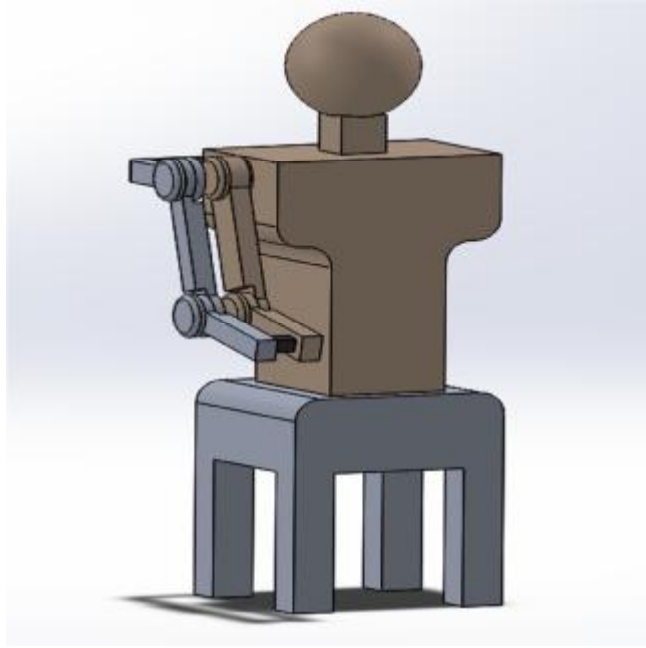


Figure 3.1. Structure of Solidworks Mechanical Design

3.1.1.1. HAMM Mechanical Models

The link lengths of HAMM are calculated using the relationship between human height and his link lengths (Winter et al., 2009). The height of patient is accepted as 175 cm in this research. Since the goal of this study is to design a controller, the possibility of human height's change is ignored during mechanical design, an orthosis and a HAMM with fixed link length are designed.

The designed orthosis has 2 degrees of freedom as stated before. In addition to this, the designed HAMM has 2 degrees of freedom. The reason of that is to implement the controller to HAMM for observing the torque changes of HAMM joints actuators during transitions of assistive/resistive rehabilitation modes.

The design of the HAMM consists of fixed part (trunk part) and moving parts (HAMM's upper arm link, HAMM's forearm link and motors). Solidworks drawing of trunk part and its dimensions can be seen in Figure 3.2. The lengths in Figures are given in centimeters (cm).

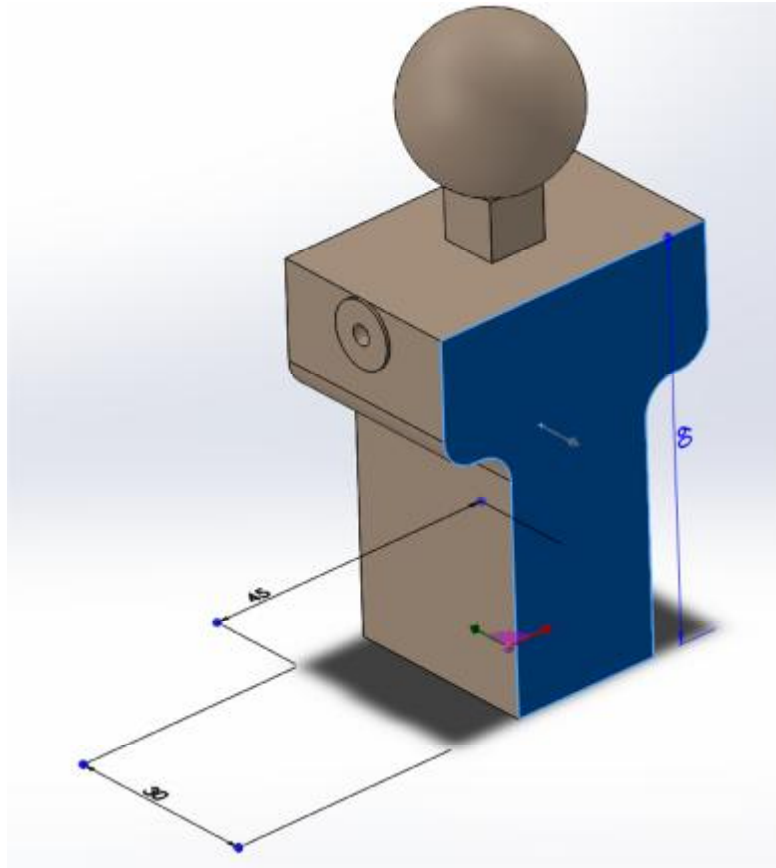
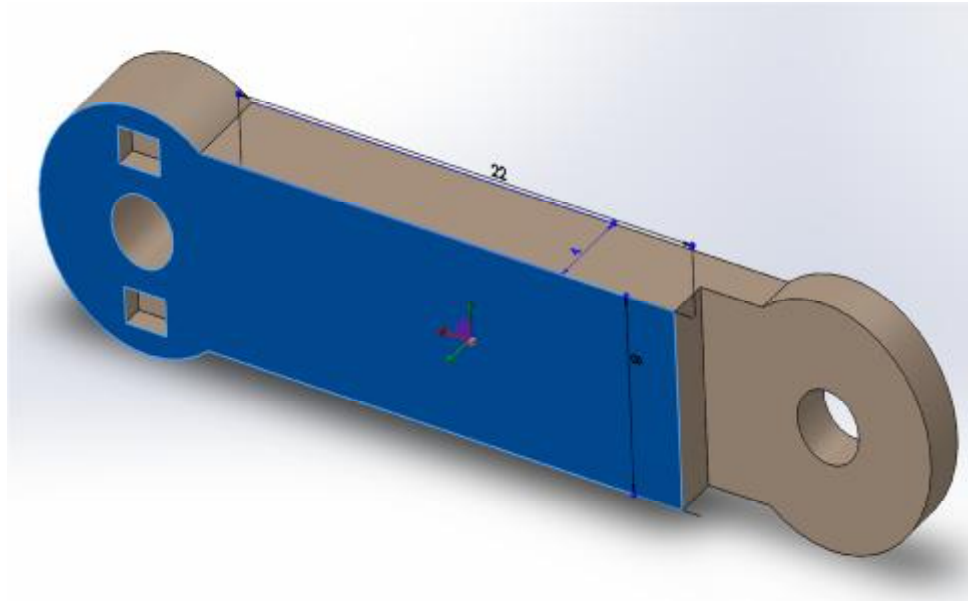
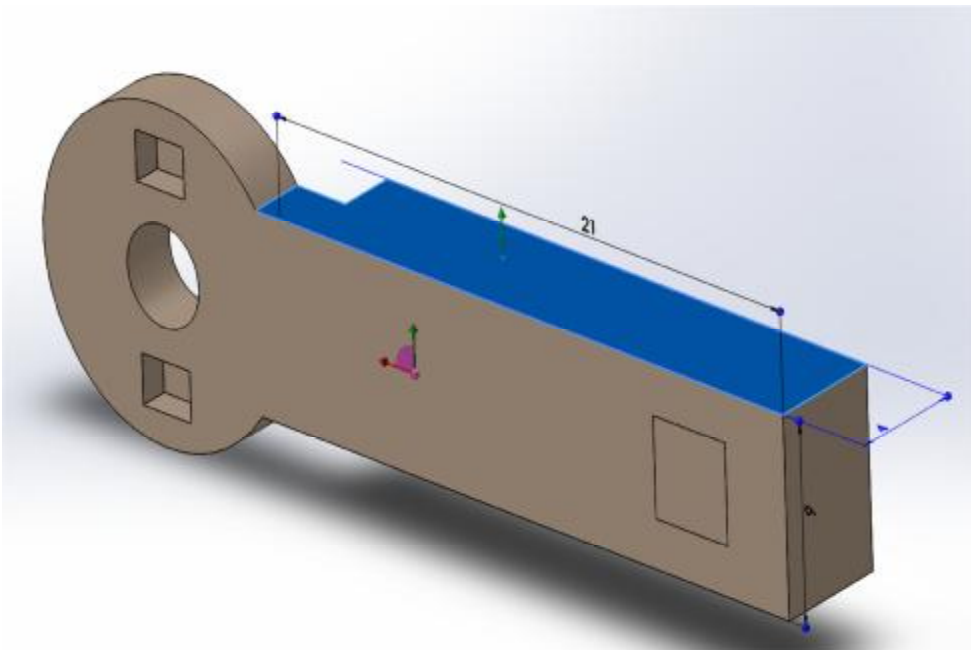


Figure 3.2. Drawing of HAMM Trunk Part

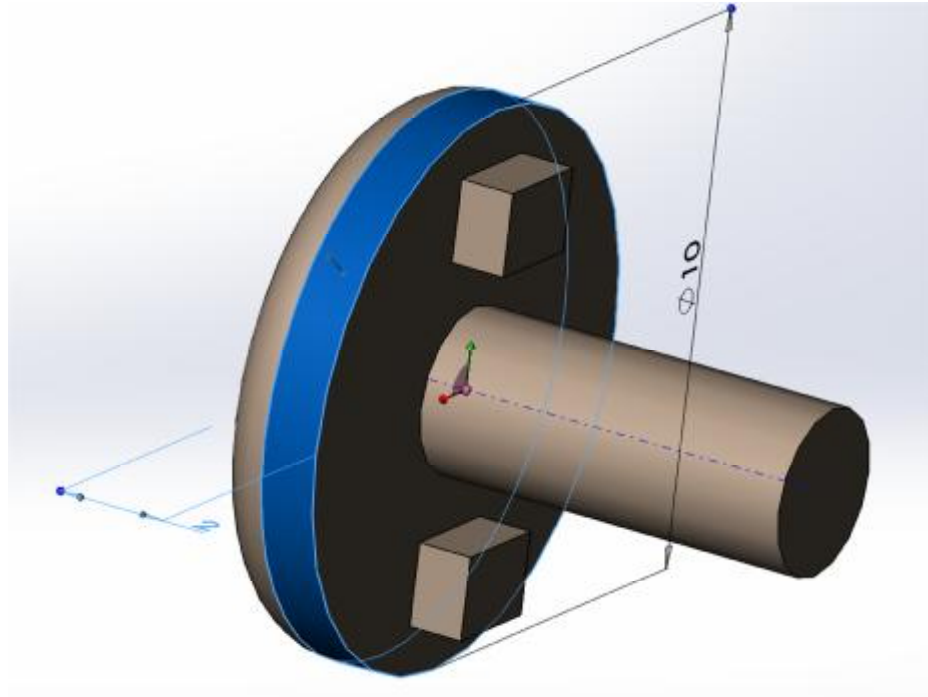
Solidworks drawing of moving parts and their dimensions can be seen in Figure 3.3.



(a) Drawing of HAMM Upperarm Part



(b) Drawing of HAMM Forearm Part

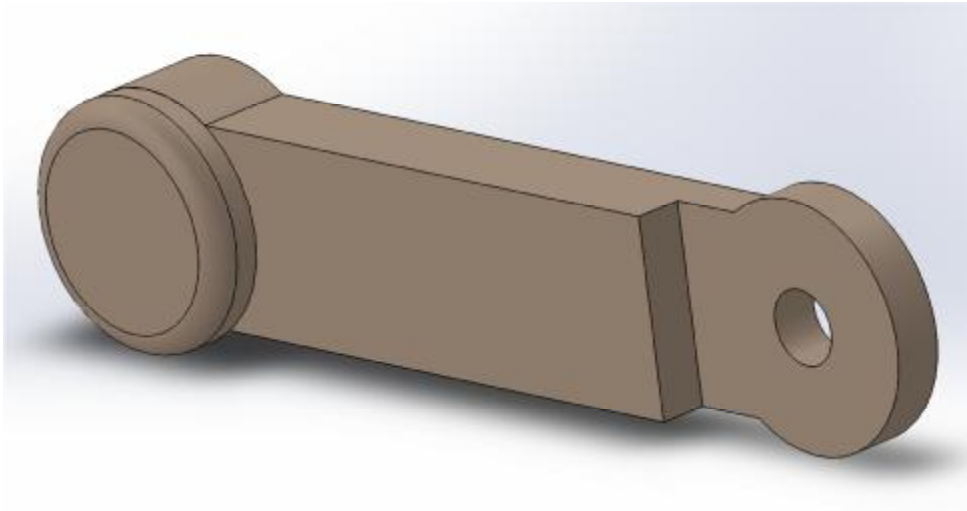


(c)

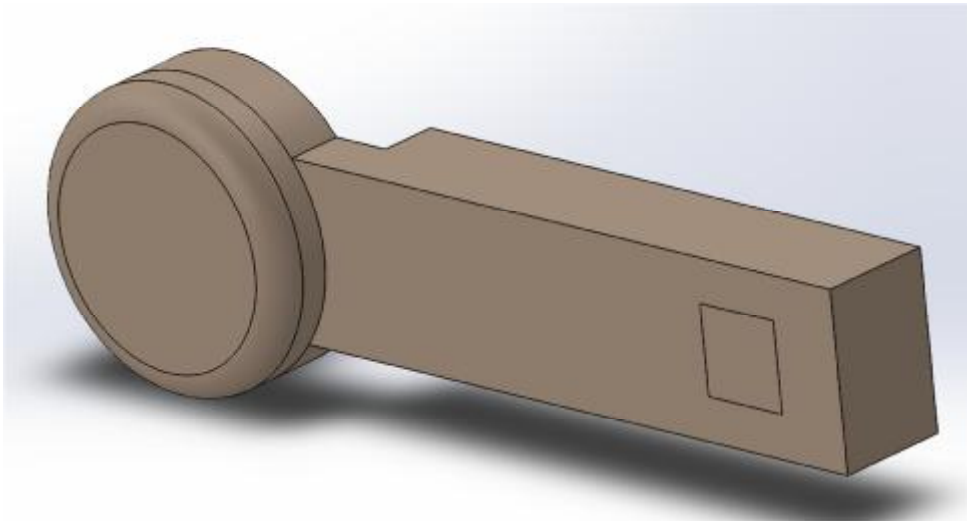
Figure 3.3. (a) Drawing of HAMM Upperarm Part (b) Drawing of HAMM Forearm Part (c) Drawing of HAMM Motors

Motors move on together with HAMM forearm and HAMM upperarm during simulation. New subassemblies are created among motors, HAMM forearm, and HAMM upperarm in Solidworks design because of that. The parameters of HAMM links like weight and moment of inertia which will be used in controllers are obtained from new subassemblies using Solidworks mass properties tool. Plain Carbon steel which exists in Solidworks material library is used as material in HAMM design to get similar values to the human arm mass values (Babiarz et al., 2014).

Solidworks appearance and Solidworks Mass Properties data of these subassemblies are shown in Figure 3.4 and Figure 3.5.

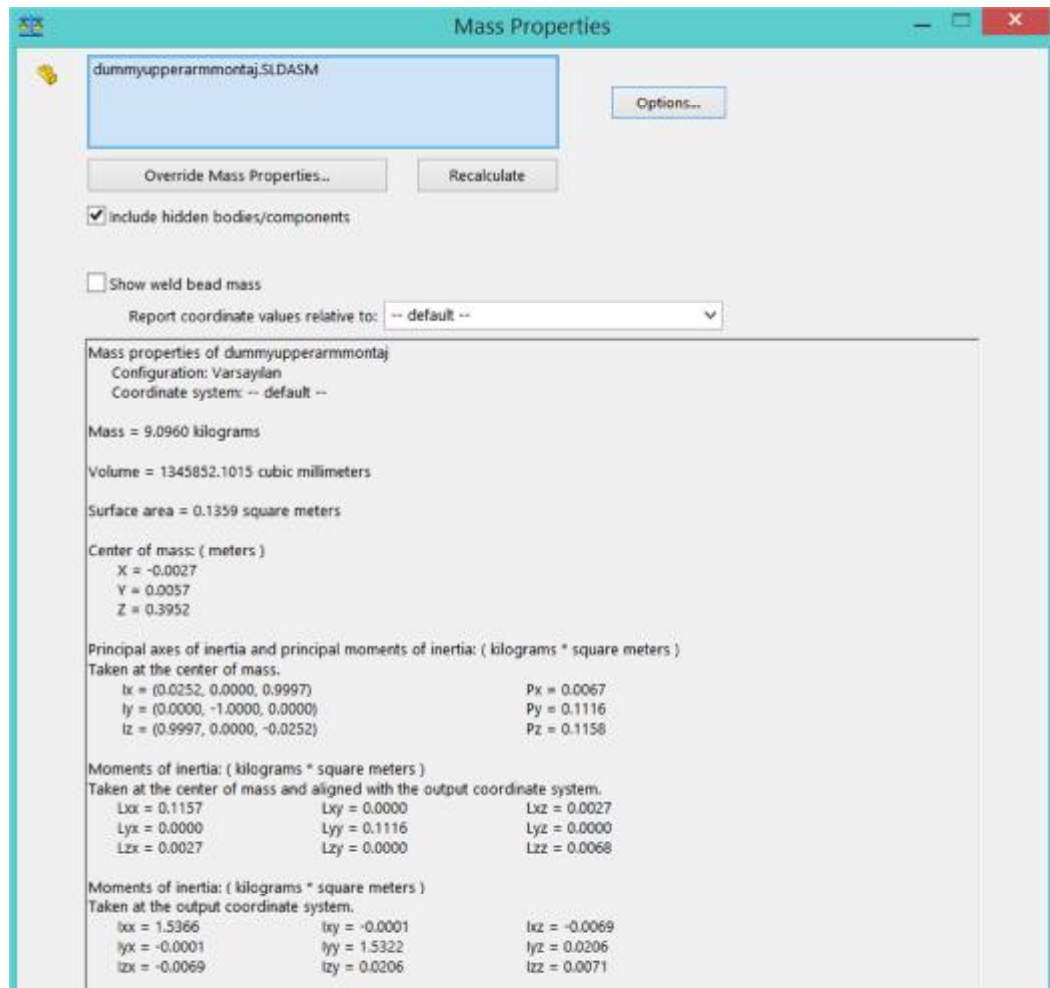


(a) Subassembly of HAMM Upperarm

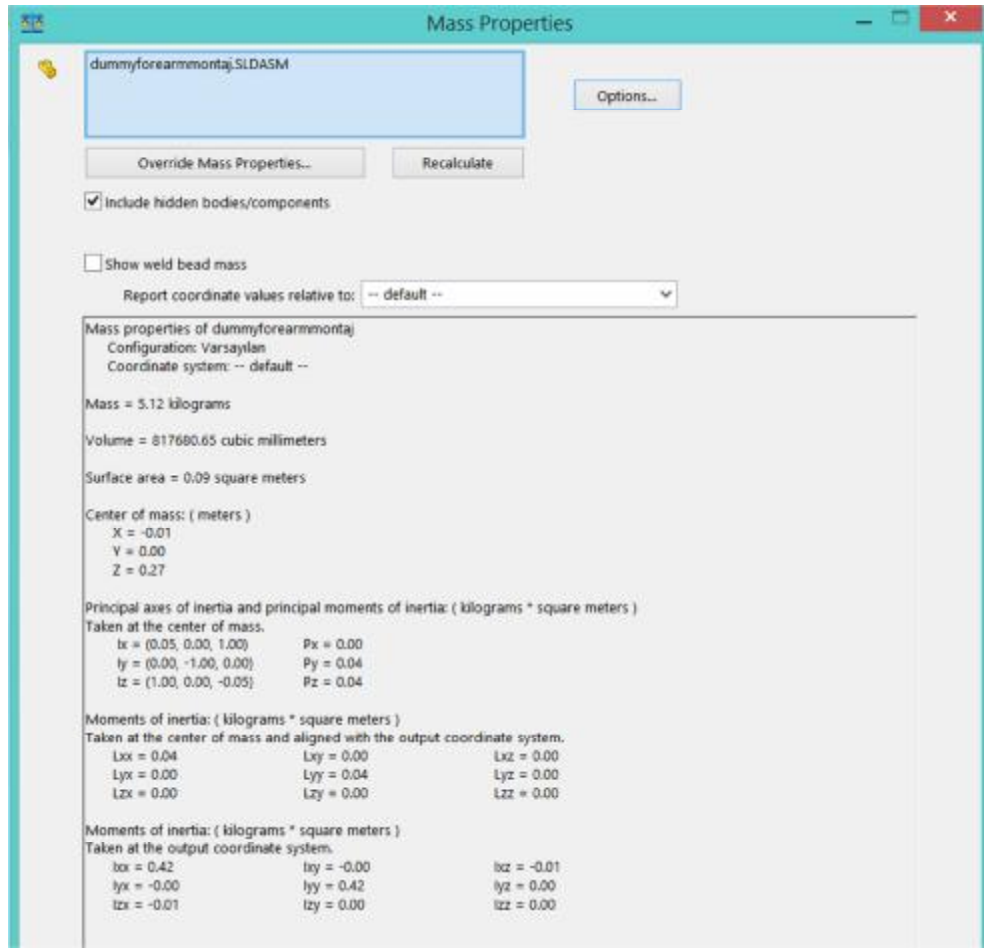


(b)

Figure 3.4. (a) Subassembly of HAMM Upperarm (b) Subassembly of HAMM Forearm



(a) Mass Properties of HAMM Upperarm Subassembly



(b)

Figure 3.5. (a) Mass Properties of HAMM Upperarm Subassembly (b) Mass Properties of HAMM Forearm Subassembly

HAMM links physical parameters which are obtained by using Solidworks tools are given in Table 3.1 where I_i represents moment of inertia, m_i is the mass, and l_{ci} the distance from the i -th joint to the i -th mass center position and l_i is the length of the i -th link.

Table 3.1. HAMM Links Physical Parameters

	m_i (Kg)	I_i (Kg m ²)	l_{ci} (m)	l_i (m)
i = 1	9.09	0.1158	0.135	0.32
I = 2	5.12	0.04	0.118	0.26

3.1.1.2. Orthosis Mechanical Models

Link lengths of orthosis are designed equally with HAMM's link lengths to prevent joint misalignments. 1060 Composite Aluminium material which exists in Solidworks library is used to make orthosis lighter.

Orthosis design like HAMM design consists of fixed part (base part) and moving parts (orthosis upperarm, orthesis forearm and motors). Solidworks drawing of base part and its dimensions are shown in Figure 3.6. The lengths in Figures are given in centimeters (cm).

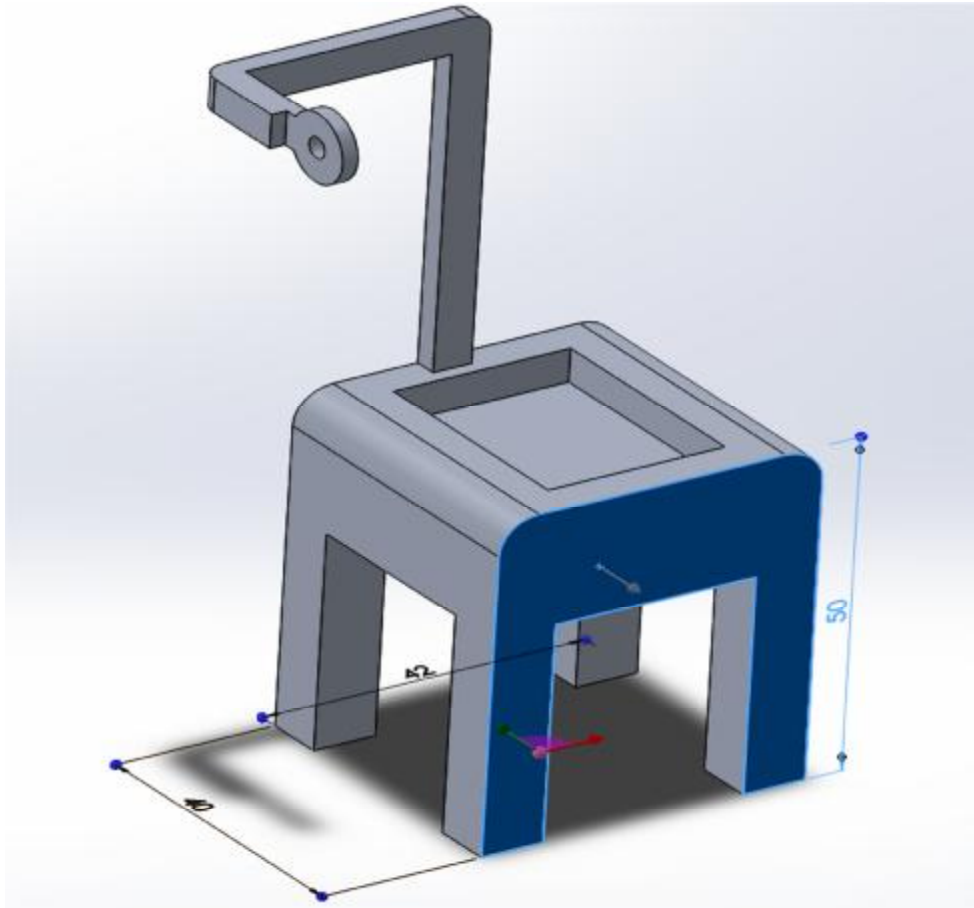
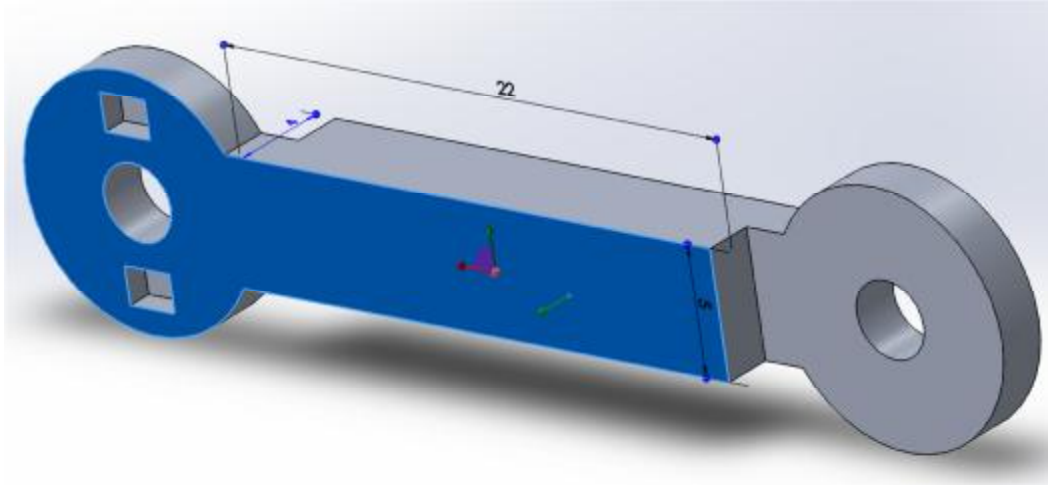
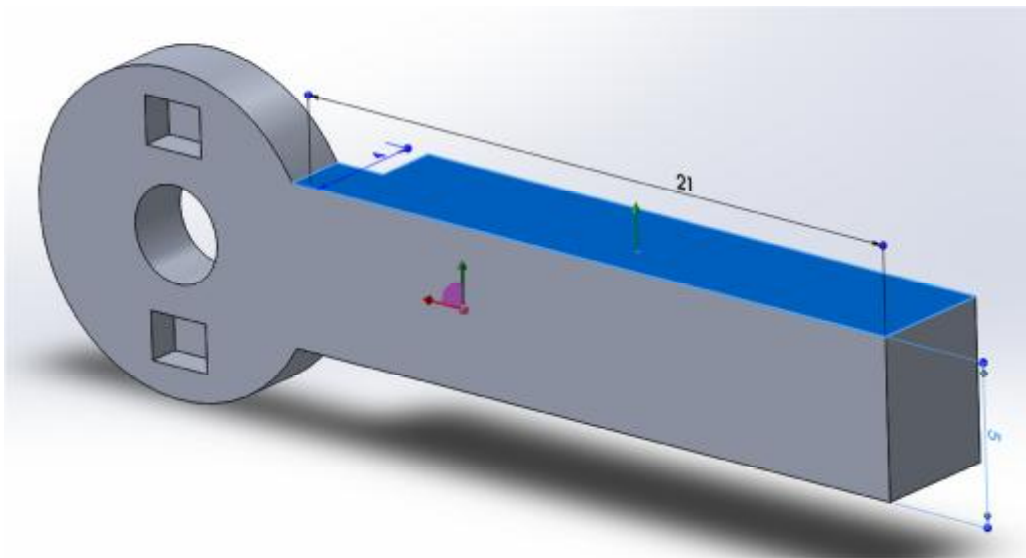


Figure 3.6. Drawing of Orthesis Base Part

Drawings of moving parts and their dimensions are shown in Figure 3.7.



(a) Drawing of Orthesis Upperarm Part



(b) Drawing of Orthesis Forearm Part

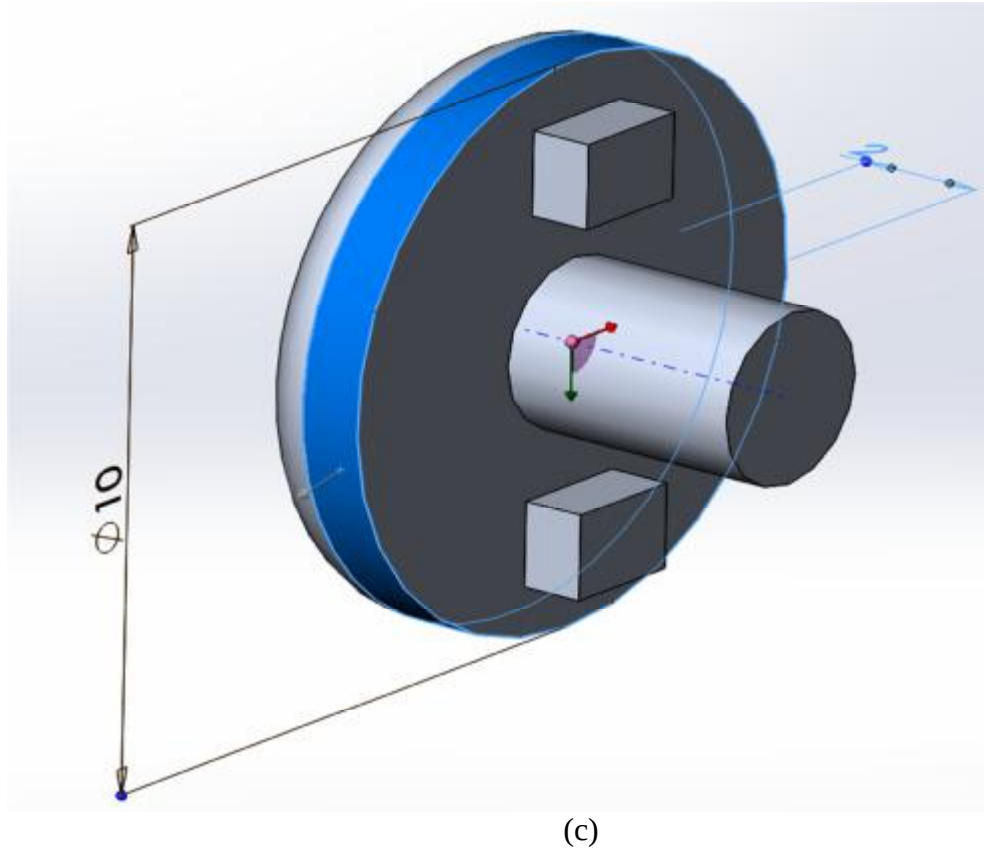
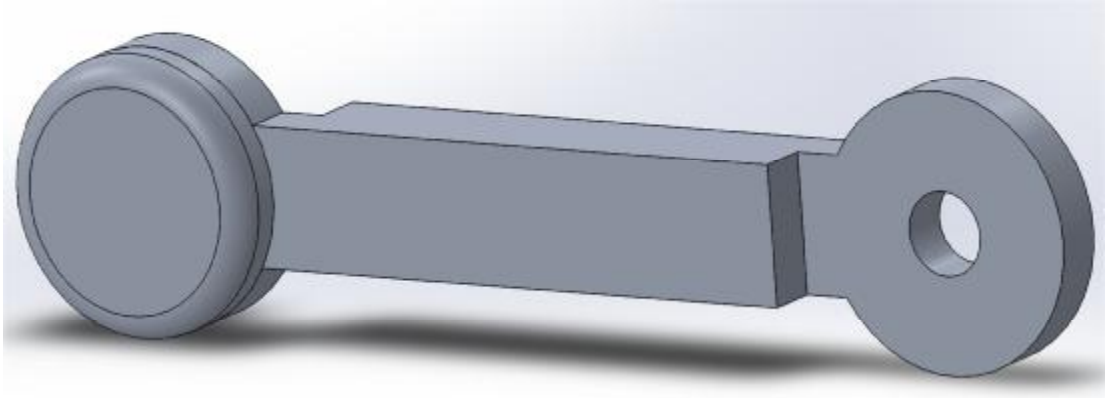


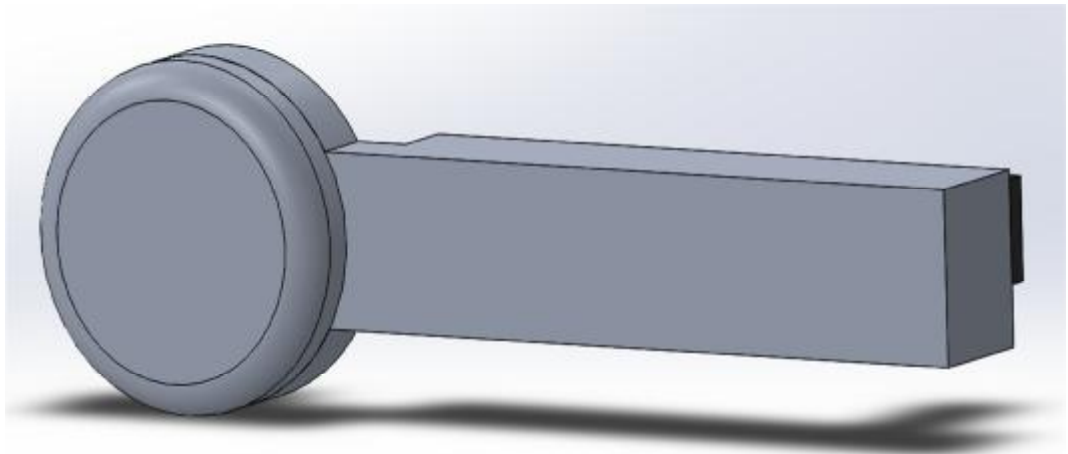
Figure 3.7. (a) Drawing of Orthosis Upperarm Part (b) Drawing of Orthesis Forearm Part (c) Drawing of Orthosis Motors

Orthosis motors move on together with orthosis forearm and orthosis upperarm during simulation as the HAMM. New subassemblies are created among motors, orthosis forearm, and orthosis upperarm in Solidworks design because of that. The parameters of orthosis links like weight and moment of inertia which will be used in controllers are obtained from new subassemblies using Solidworks mass properties tool.

Solidworks appearance and Solidworks Mass Properties data of these subassemblies are shown in Figure 3.8 and Figure 3.9.

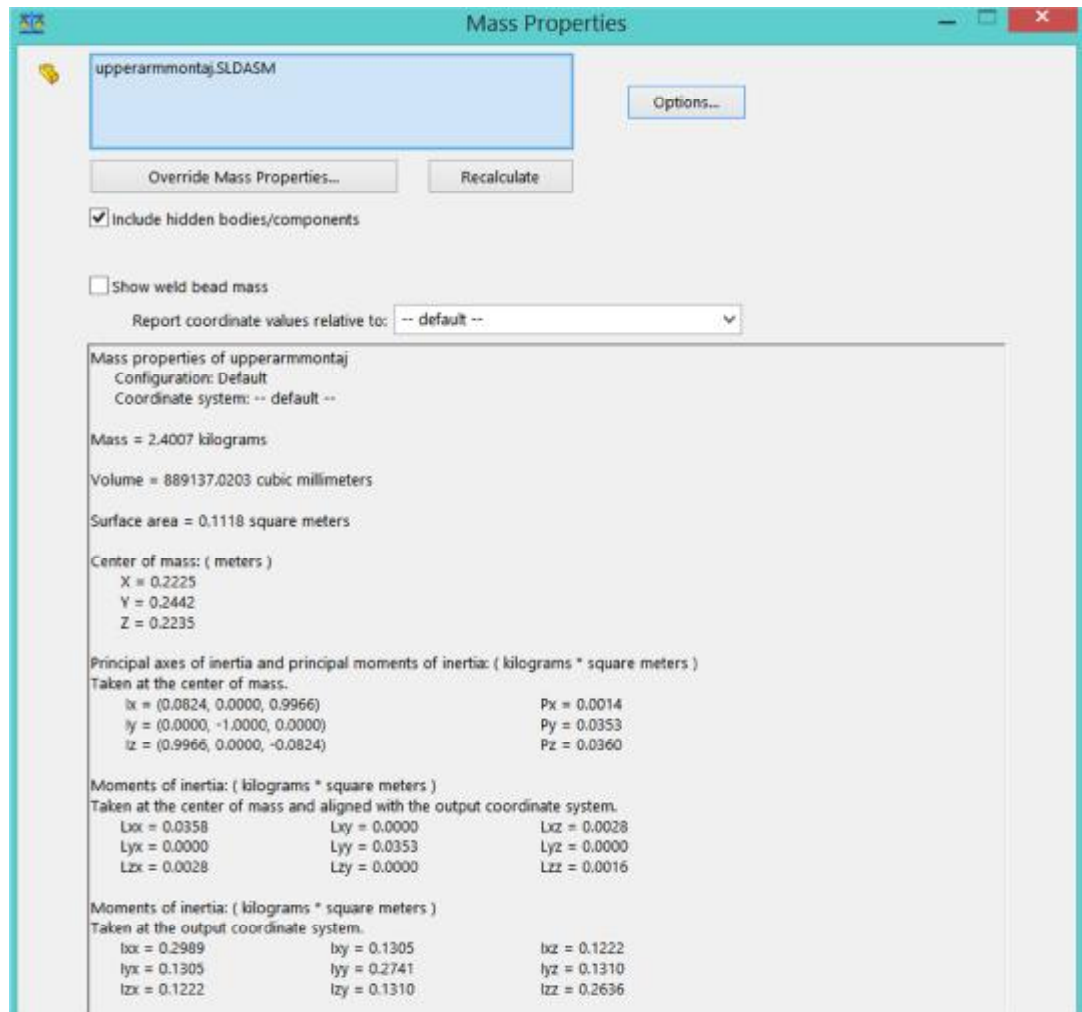


(a) Subassembly of Orthosis Upperarm

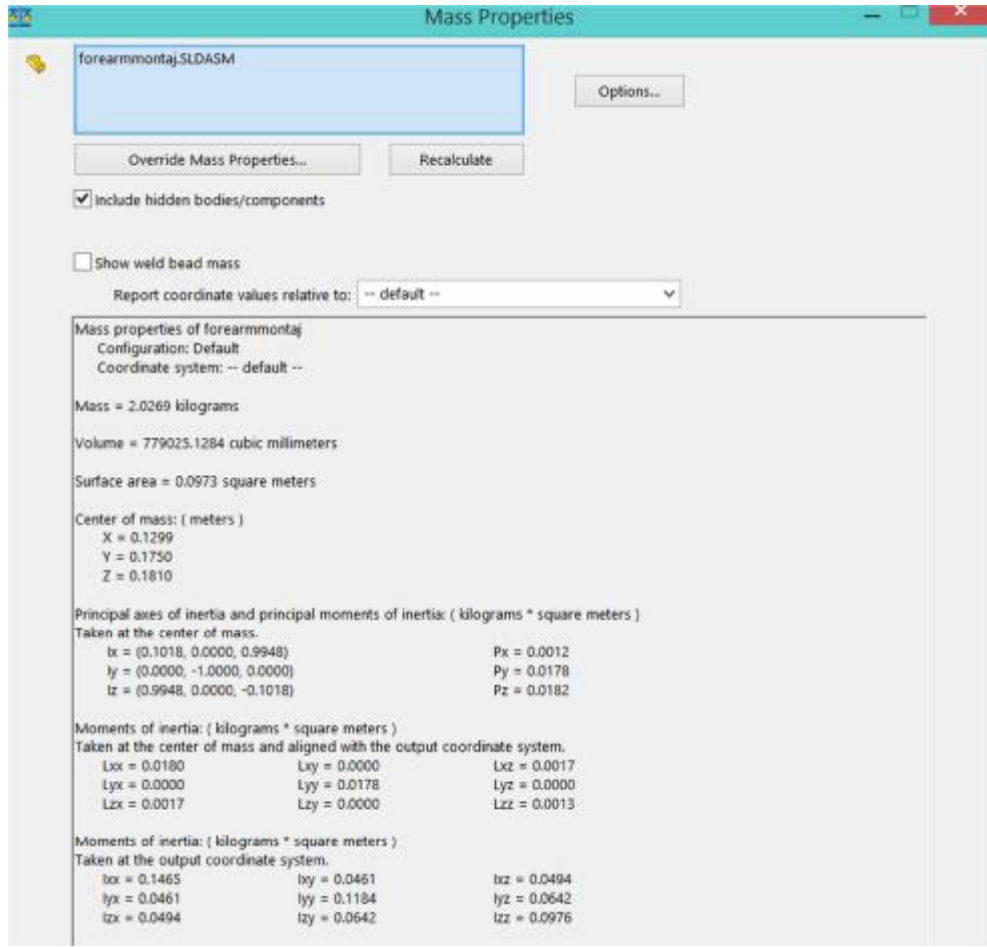


(b)

Figure 3.8. (a) Subassembly of Orthosis Upperarm (b) Subassembly of Orthosis Forearm



(a) Mass Properties of Orthosis Upperarm Subassembly



(b)

Figure 3.9. (a) Mass Properties of Orthosis Upperarm Subassembly (b) Mass Properties of Orthosis Forearm Subassembly

Orthosis links physical parameters which are obtained by using Solidworks tools are given in Table 3.2 where I_i represents moment of inertia, m_i is the mass, and l_{ci} the distance from the i -th joint to the i -th mass center position and l_i is the length of the i -th link.

Table 3.2. Orthosis Links Physical Parameters

	m_i (Kg)	I_i (Kg m ²)	l_{ci} (m)	l_i (m)
i = 1	2.4	0.036	0.128	0.32
I = 2	2.02	0.0182	0.099	0.26

3.1.2. SimMechanics Link Software Between Solidworks and Matlab

MATLAB programming language is used for designing orthosis controller. A mechanical model of orthosis and HAMM are needed in Matlab to implement the designed controller. These models are obtained from transferring Solidworks designs of orthosis and HAMM to Matlab using Matlab Simmechanics software (Figure 3.10). The SimMechanics Link software provides the primary interface for exporting CAD assemblies into Matlab/SimMechanics models².

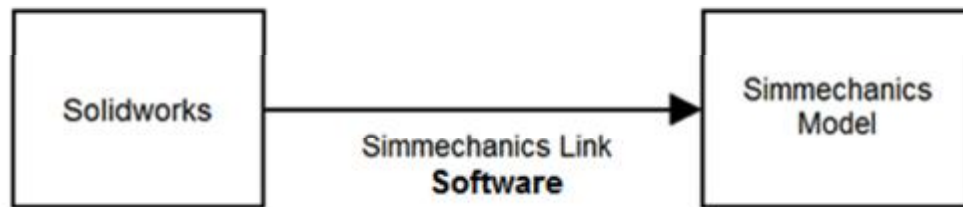


Figure 3.10. Simmechanics Link Software

Simmechanics Link software was downloaded directly from MathWorks website³ and installed into the Matlab. After the installation process, Simmechanics Link plug-in must be enabled on Solidworks application. For this, “smlink_linksw” command was entered to the Matlab command window. Then, the installation process was completed by selecting the SimMechanics Link check box from Solidworks add-ins menu.

Files transferred by using Simmechanics Link plug-in from Solidworks application are shown in Figure 3.11.

²<http://www.mathworks.com/help/physmod/smlink/ref/linking-and-unlinking-simmechanics-link-software-with-solidworks.html>

³https://www.mathworks.com/products/simmechanics/download_smlink.html



Figure 3.11. Files Obtained by Simmechanics Link Software

STL spurred files in Figure 3.11 include visual details of design, XML spurred files include data such as mass values, coordination of mass center, inertia matrix data. “mech_import (‘filename.xml’)” command in Matlab command window is used to get the mechanical models of orthesis and HAMM in Matlab with Simulink/Simmechanics First Generation blocks by using data from XML file. The model obtained from using this command is shown in Figure 3.12. The mechanical model designed in Solidworks is exported to Matlab as a result of all these processes.

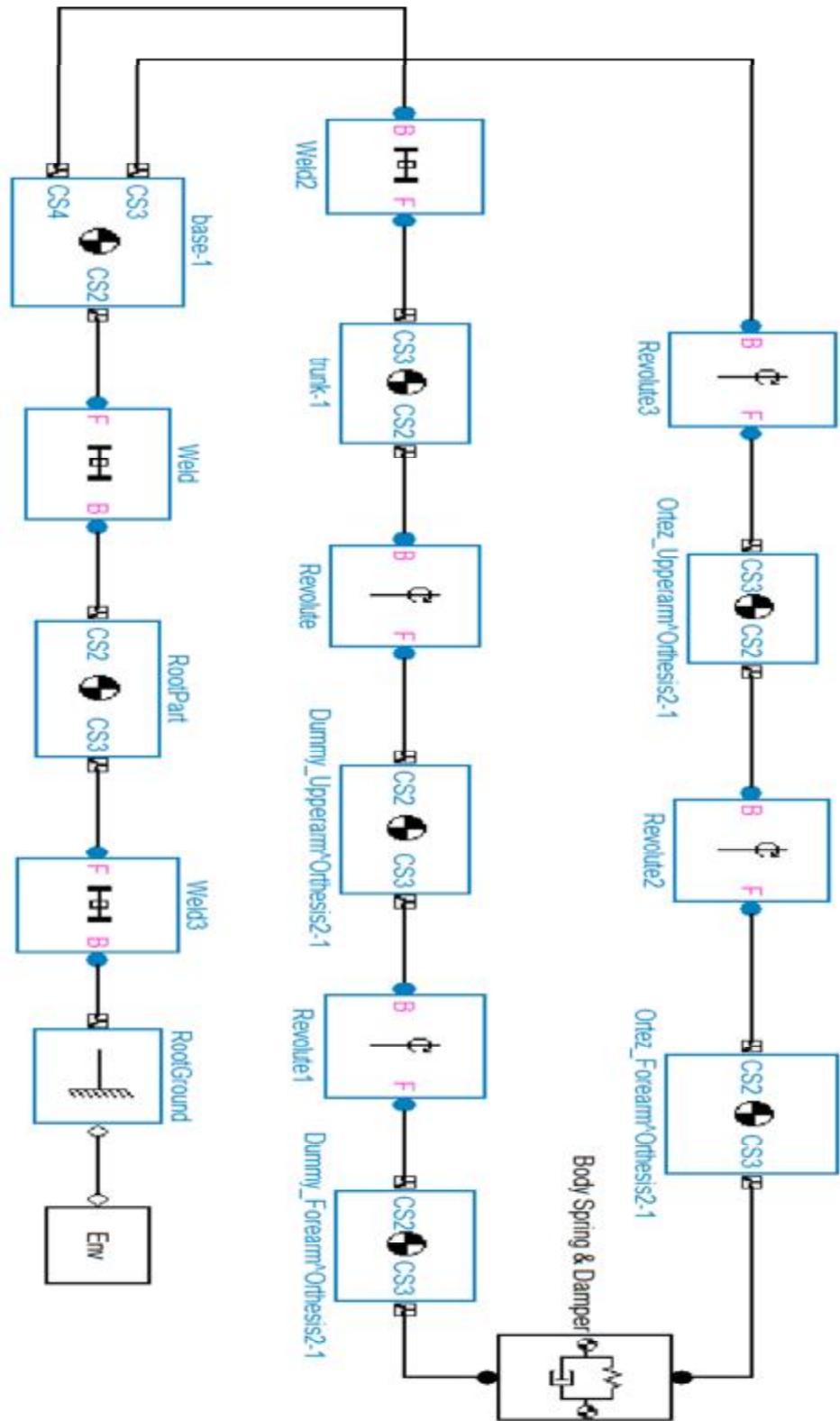


Figure 3.12. Transferred Mechanical Model of Orthosis and HAMM to Matlab

3.1.3. Matlab Model

The structure of controller and mechanical models of HAMM and orthesis are designed in Matlab with Matlab/Simulink and Simmechanics First Generation blocks.

Simulink is a block diagram environment for multidomain simulation and Model-Based Design. It supports simulation, automatic code generation, and continuous test and verification of embedded systems. It is integrated with Matlab, enabling you to incorporate Matlab algorithms into models and export simulation results to Matlab for further analysis.

SimMechanics provides a multibody simulation environment for 3D mechanical systems, such as robots, vehicle suspensions, construction equipment, and aircraft landing gear. User model the multibody system using blocks representing bodies, joints, constraints, and force elements, and then SimMechanics formulates and solves the equations of motion for the complete mechanical system. Models from CAD systems, including mass, inertia, joint, constraint, and 3D geometry, can be imported into SimMechanics. An automatically generated 3D animation lets the user visualize the system dynamics.

The model created in Matlab is shown in Figure 3.13 and Figure 3.14.

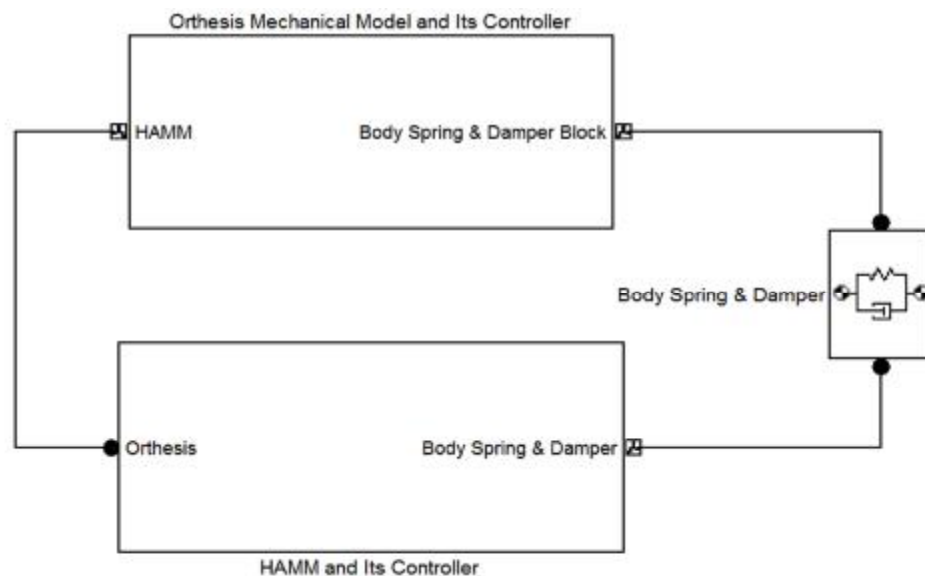
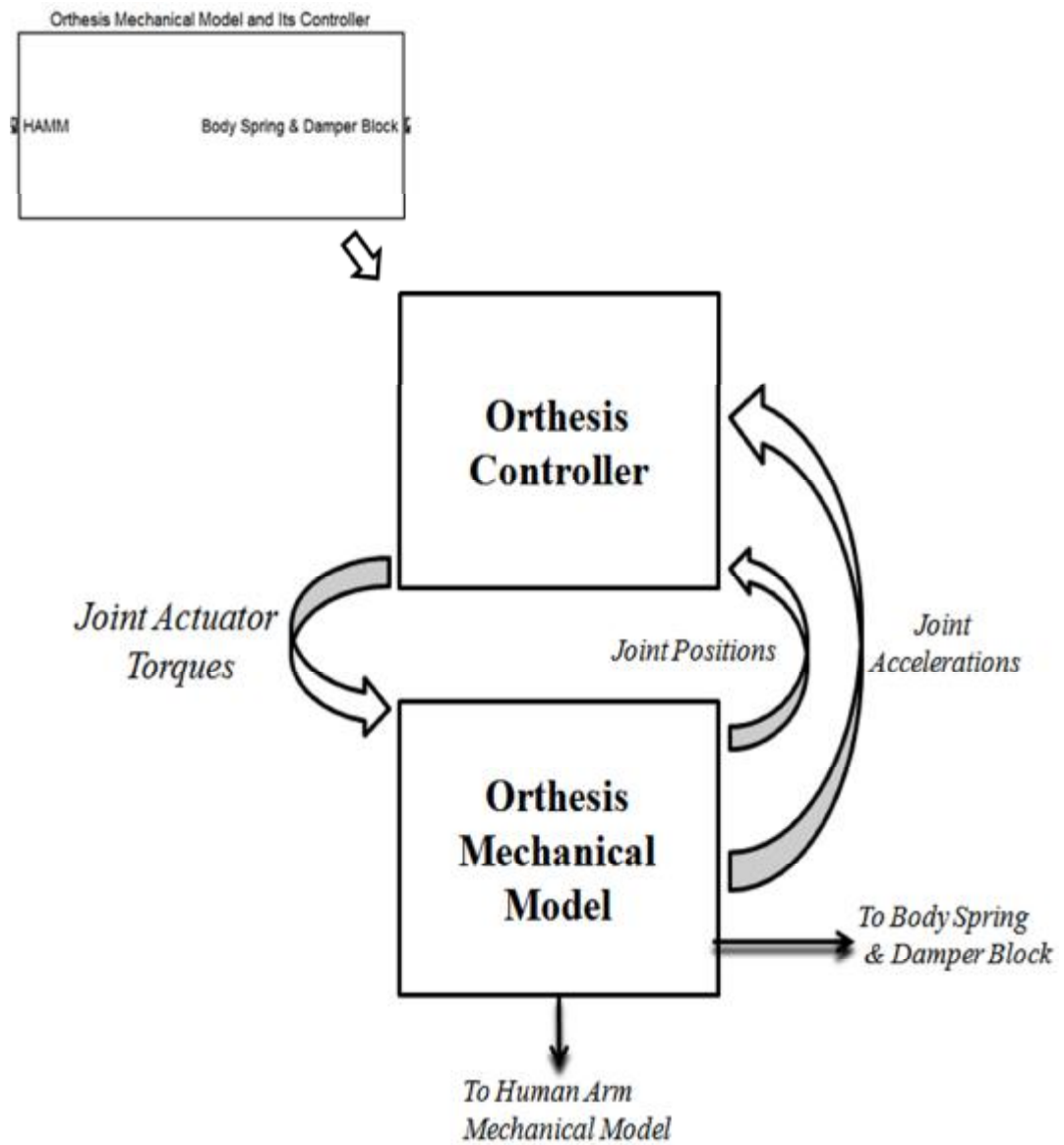
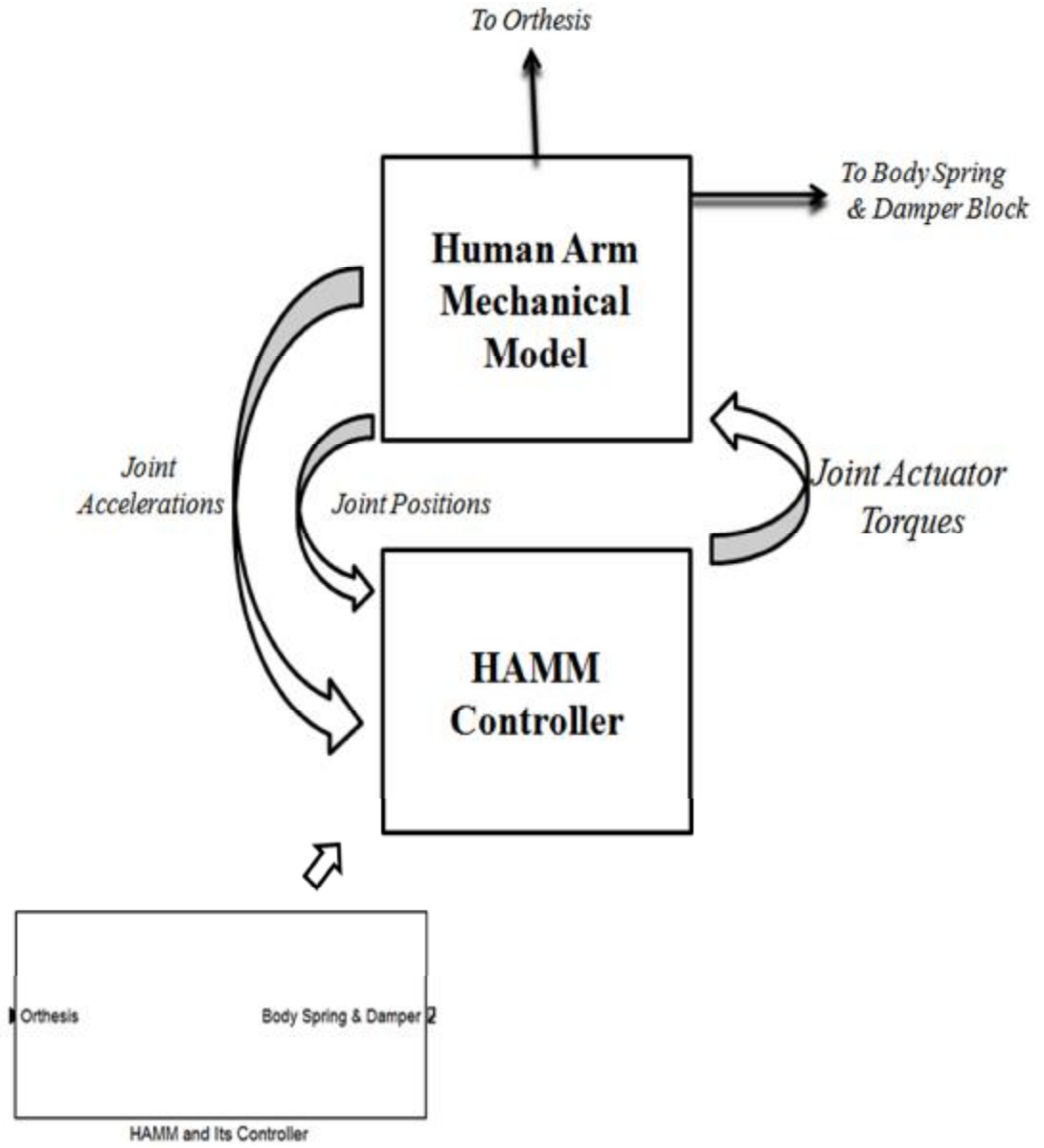


Figure 3.13. Matlab/Simulink Model of the System



(a) Matlab Model of Orthosis and Its Controller



(b)

Figure 3.14. (a) Matlab Model of Orthosis and Its Controller (b) Matlab Model of HAMM and Its Controller

The blocks which make the HAMM and orthosis are explained in this section. The controller structure is explained in the next section.

3.1.3.1. Simmechanics Mechanical Model

Mechanical models of HAMM and orthosis consist of Simmechanics First Generation body and joint blocks. Blocks, block parameters, body geometry and color data are obtained from Solidworks CAD drawings by using Simmechanics Link software.

Simmechanics body block represents user defined rigid body (Figure 3.15). Body block parameters are shown in Figure 3.16.



Figure 3.15. Simmechanics Body Block

Body

Represents a user-defined rigid body. Body defined by mass m , inertia tensor I , and coordinate origins and axes for center of gravity (CG) and other user-specified Body coordinate systems. This dialog sets Body initial position and orientation, unless Body and/or connected Joints are actuated separately. This dialog also provides optional settings for customized body geometry and color.

Mass properties

Mass: 1 kg

Inertia: eye(3) kg*m^2

Position | Orientation | Visualization

Show Port	Port Side	Name	Origin Position Vector [x y z]	Units	Translated from Origin of	Components in Axes of
☐	Left	CG	[0 0 0]	m	World	World
☒	Left	CS1	[0 0 0]	m	CG	CG
☒	Right	CS2	[0 0 0]	m	CG	CG

OK Cancel Help Apply

Figure 3.16. Body Block Parameters

This body block consists of mass, inertia tensor, coordinates of Center of Gravity and user-specified coordinate systems. The inertia is defined with a 3x3

matrix. Coordinate Systems (CS) represent the coordinates which connects body with neighboring body with Joint, Constraint, or Driver.

HAMM mechanical model consists of 3 body blocks such as fixed body block (trunk block) and moving body blocks which are Dummy_upperarm and Dummy_forearm (Figure 3.17).

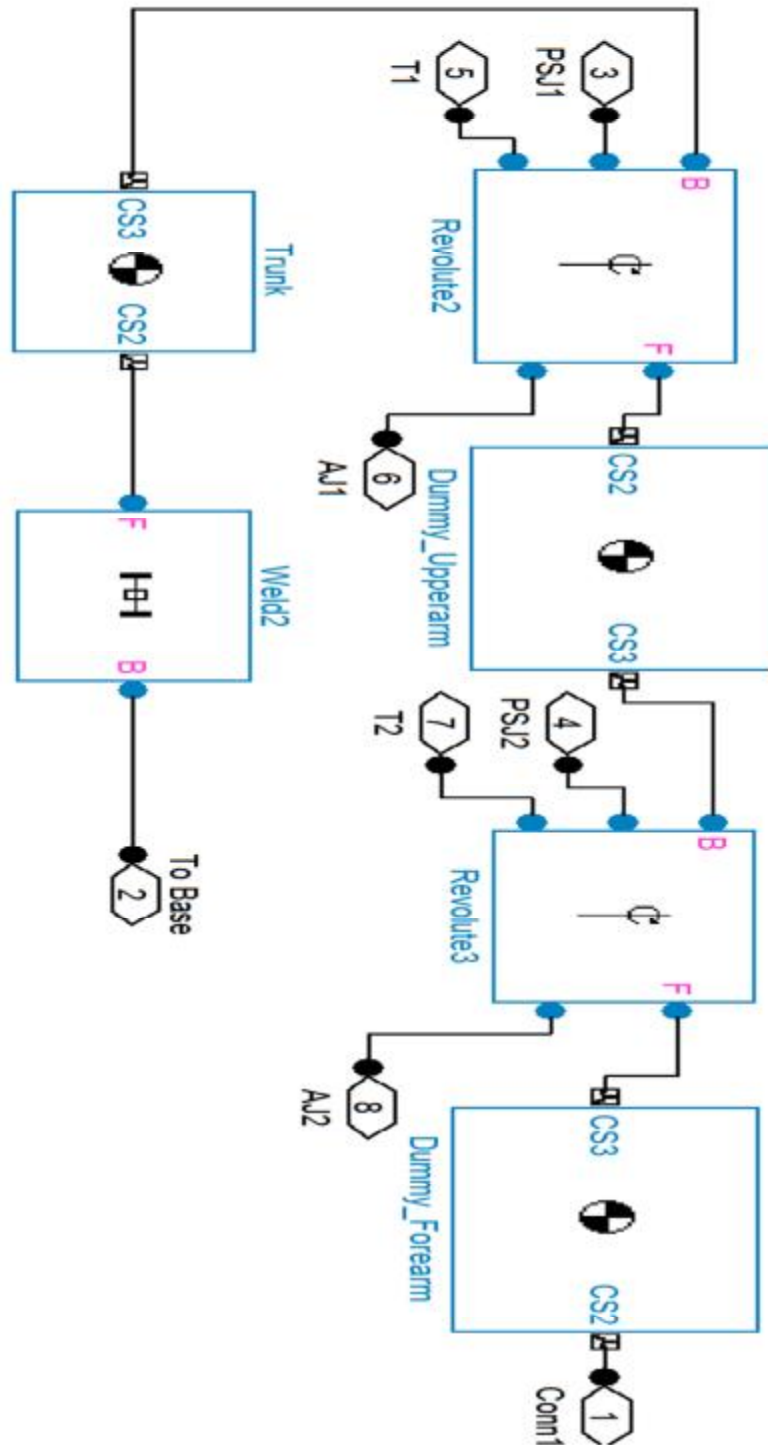


Figure 3.17. HAMM Mechanical Model

HAMM trunk block parameters are shown in Figure 3.18.

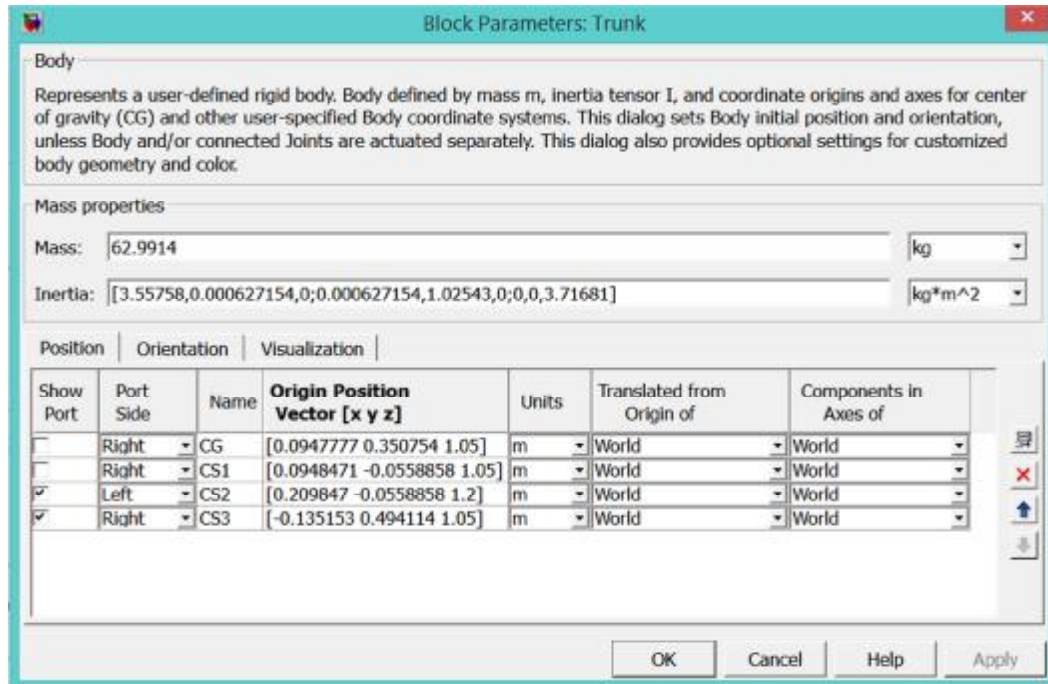


Figure 3.18. HAMM Trunk Block Parameters

This block is connected to dummy_upperarm block from CS3 point with revolute joint, it is also connected to orthosis body block from CS2 point with weld joint as shown in Figure 3.17.

Weld joint represents zero degrees of freedom (Figure 3.19). This joint rigidly connects Base (B) and Follower (F). Weld joint block parameters are shown in Figure 3.20. Sensor ports can be added to this joint. Weld joint cannot be actuated.

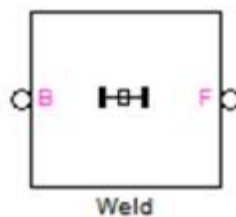


Figure 3.19. Weld Joint Block

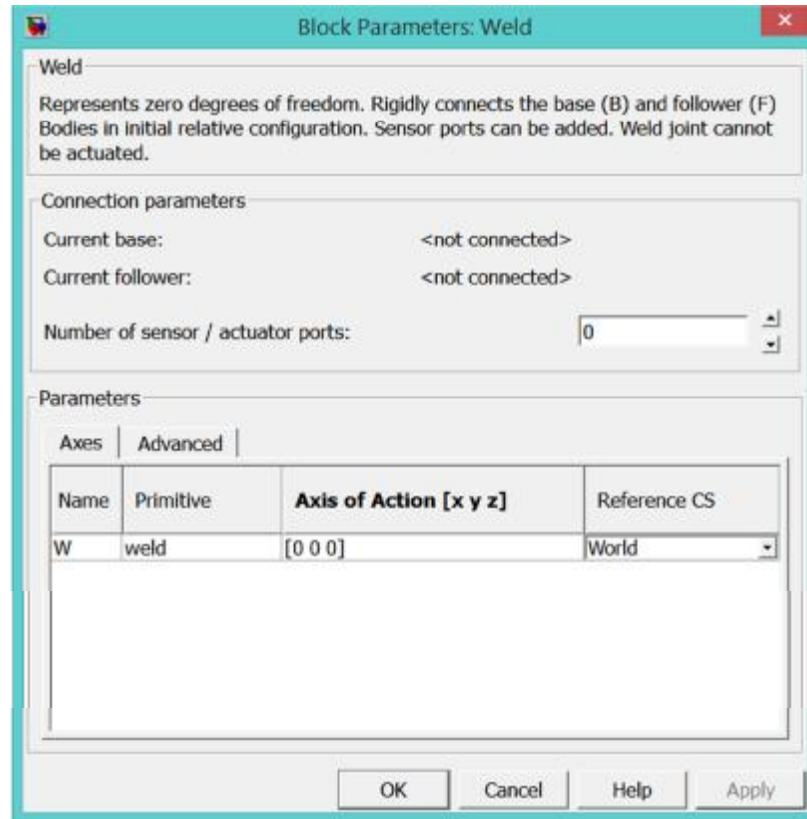


Figure 3.20. Weld Joint Block Parameters

Revolute joint represents one rotational degree of freedom (Figure 3.21). Revolute joint which is used in the model, block parameters is shown in Figure 3.22. The follower (F) Body of this joint rotates relative to the base (B) Body. Sensor and actuator ports can be added to this joint. Base-follower sequence and axis direction determine sign of forward motion by the right-hand rule.

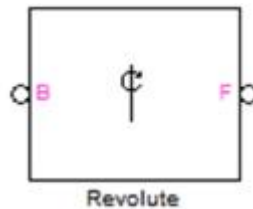


Figure 3.21. Revolute Joint Block Parameters

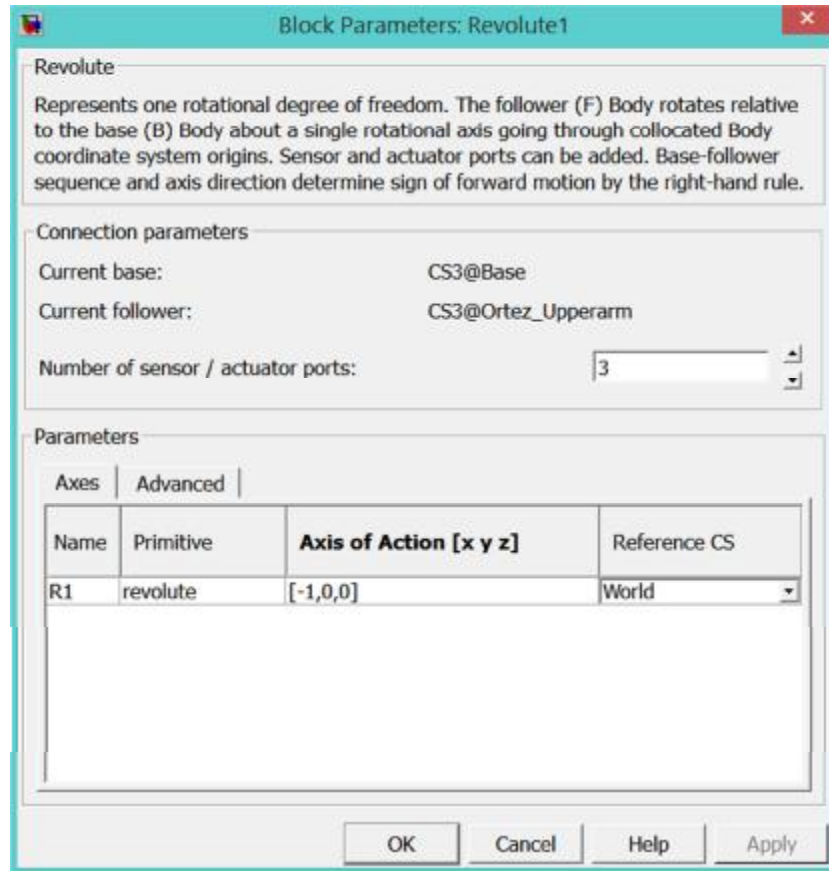


Figure 3.22. Revolute Joints Block Parameters

Axis of Action field represents which direction(s) the motion is to take place. An Axis of Action of $[-1\ 0\ 0]$ tells the motion take place in $\pm x$ direction. Meaning of minus sign in front of 1 is negative movement in x direction will be considered as positive. All revolute joints in this model have the same Axis of Action.

Block parameters of Dummy_upperarm and Dummy_forearm which are moving blocks of HAMM mechanical design are shown in Figure 3.23.

Block Parameters: Dummy_Upperarm

Body
Represents a user-defined rigid body. Body defined by mass m , inertia tensor I , and coordinate origins and axes for center of gravity (CG) and other user-specified Body coordinate systems. This dialog sets Body initial position and orientation, unless Body and/or connected Joints are actuated separately. This dialog also provides optional settings for customized body geometry and color.

Mass properties

Mass: 9.09604 kg

Inertia: [0.115685, 0, -0.00274932; 0, 0.111613, 0; -0.00274932, 0, 0.00675224] kg*m²

Position | Orientation | Visualization

Show Port	Port Side	Name	Origin Position Vector [x y z]	Units	Translated from Origin of	Components in Axes of
☐	Right	CG	[-0.15405 0.358925 1.05]	m	World	World
☐	Right	CS1	[-0.175153 0.334114 1.05]	m	World	World
☒	Left	CS2	[-0.135153 0.494114 1.05]	m	World	World
☒	Right	CS3	[-0.135153 0.174114 1.05]	m	World	World

OK Cancel Help Apply

(a)

Block Parameters: Dummy_Forearm

Body
Represents a user-defined rigid body. Body defined by mass m , inertia tensor I , and coordinate origins and axes for center of gravity (CG) and other user-specified Body coordinate systems. This dialog sets Body initial position and orientation, unless Body and/or connected Joints are actuated separately. This dialog also provides optional settings for customized body geometry and color.

Mass properties

Mass: 5.1205 kg

Inertia: [0.0398806, 0, -0.00185531; 0, 0.0386494, 0; -0.00185531, 0, 0.00272014] kg*m²

Position | Orientation | Visualization

Show Port	Port Side	Name	Origin Position Vector [x y z]	Units	Translated from Origin of	Components in Axes of
☐	Right	CG	[-0.158325 0.055846 1.05]	m	World	World
☐	Right	CS1	[-0.175153 0.0191142 1.05]	m	World	World
☒	Right	CS2	[-0.175153 -0.0308858 1.065]	m	World	World
☒	Left	CS3	[-0.135153 0.174114 1.05]	m	World	World

OK Cancel Help Apply

(b)

Figure 3.23. (a) Dummy_upperarm Block Parameters (b) Dummy_forearm Block Parameters

Dummy_forearm block is connected to orthesis_forearm block with body spring damper block (Figure 3.24). Body spring damper block parameters are shown in Figure 3.25.

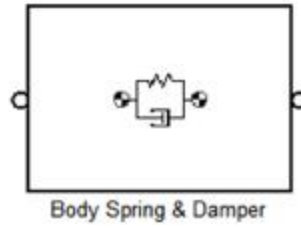


Figure 3.24. Body Spring & Damper Block

Block Parameters: Body Spring & Damper

Body Spring & Damper
Models a damped linear oscillator between two Bodies, equivalent to a translational spring and damper. The force F between the bodies is projected along the axis connecting the Body coordinate systems and is a function of the relative displacement r and velocity v of these Body coordinate systems, given by $F = -k*(r-r_0) - b*v$. The parameters r_0 , k , and b represent the spring's natural length, the spring constant, and the damper constant, respectively.

Parameters

Spring constant (k): 95000

Damper constant (b): 500

Spring natural length (r_0): 0.001

Units

Position: m

Velocity: m/s

Force: N

OK Cancel Help Apply

Figure 3.25. Body Spring & Damper Block Parameters

This block models a damped linear oscillator between two Bodies, equivalent to a translational spring and damper. Within the block the user can define spring constant, damper constant and initial length of the spring values where the spring and damper are in paralel.

This block is used to connect HAMM and orthosis in a flexible manner to simulate a real connection. It is also aimed to observe the interaction between HAMM and orthosis during mode transitions in controller's structure.

The mechanical model of orthosis consists of 3 body blocks as shown in Figure 3.26. Base block of these is fixed, the orthosis_upperarm and orthosis_forearm are moving blocks.

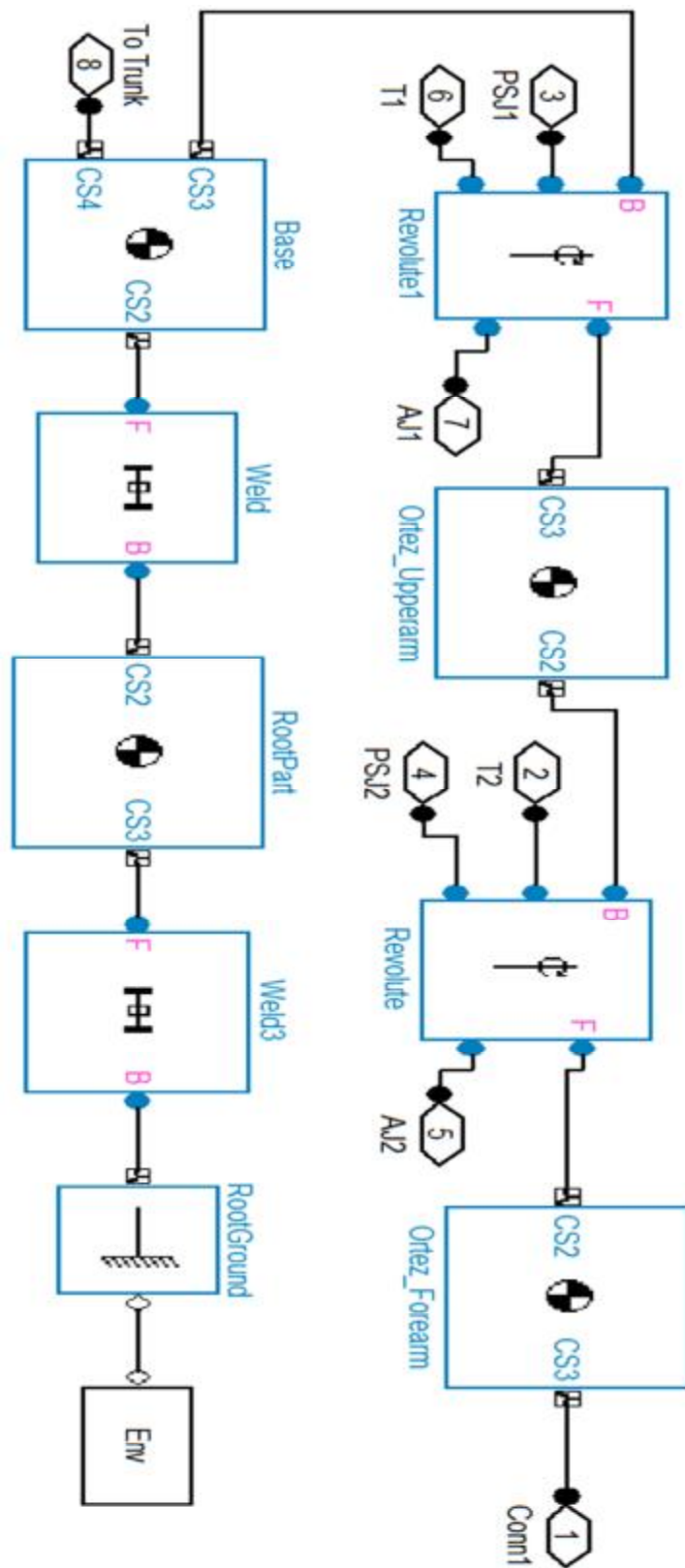


Figure 3.26. Orthesis Mechanical Model

Block parameters of these blocks are shown in Figure 3.25.

Block Parameters: Base

Body
Represents a user-defined rigid body. Body defined by mass m , inertia tensor I , and coordinate origins and axes for center of gravity (CG) and other user-specified Body coordinate systems. This dialog sets Body initial position and orientation, unless Body and/or connected Joints are actuated separately. This dialog also provides optional settings for customized body geometry and color.

Mass properties
Mass: 94.319 kg
Inertia: [4.2346, 0.390594, -0.0747805; 0.390594, 3.52846, 0.338906; -0.0747805, 0.338906, 4.44836] kg*m²

Position | Orientation | Visualization

Show Port	Port Side	Name	Origin Position Vector [x y z]	Units	Translated from Origin of	Components in Axes of
☐	Right	CG	[0.088047 -0.134561 1.04271]	m	World	World
☐	Right	CS1	[0.0948471 -0.505886 1.05]	m	World	World
☒	Left	CS2	[0.088047 -0.134561 1.04271]	m	World	World
☒	Right	CS3	[-0.245153 0.494114 1.05]	m	World	World
☒	Right	CS4	[0.209847 -0.0558858 1.2]	m	World	World

OK Cancel Help Apply

(a) Orthosis Base Block Parameters

Block Parameters: Ortez_Upperarm

Body
Represents a user-defined rigid body. Body defined by mass m , inertia tensor I , and coordinate origins and axes for center of gravity (CG) and other user-specified Body coordinate systems. This dialog sets Body initial position and orientation, unless Body and/or connected Joints are actuated separately. This dialog also provides optional settings for customized body geometry and color.

Mass properties
Mass: 2.40067 kg
Inertia: [0.0357724, 0, -0.00284712; 0, 0.0353327, 0; -0.00284712, 0, 0.0015922] kg*m²

Position | Orientation | Visualization

Show Port	Port Side	Name	Origin Position Vector [x y z]	Units	Translated from Origin of	Components in Axes of
☐	Right	CG	[-0.25018 0.36631 1.05]	m	World	World
☐	Right	CS1	[-0.265153 0.334114 1.05]	m	World	World
☒	Right	CS2	[-0.265153 0.174114 1.05]	m	World	World
☒	Left	CS3	[-0.245153 0.494114 1.05]	m	World	World

OK Cancel Help Apply

(b) Orthosis_upperarm Block Parameters

Block Parameters: Ortez_Forearm

Body
Represents a user-defined rigid body. Body defined by mass m , inertia tensor I , and coordinate origins and axes for center of gravity (CG) and other user-specified Body coordinate systems. This dialog sets Body initial position and orientation, unless Body and/or connected Joints are actuated separately. This dialog also provides optional settings for customized body geometry and color.

Mass properties

Mass: 2.02687 kg

Inertia: [0.0179974,0,-0.00172296;0,0.0178304,0;-0.00172296,0,0.00133022] kg*m²

Position | Orientation | Visualization

Show Port	Port Side	Name	Origin Position Vector [x y z]	Units	Translated from Origin of	Components in Axes of
☐	Right	CG	[-0.252174 0.0831037 1.05]	m	World	World
☐	Right	CS1	[-0.265153 0.0191142 1.05]	m	World	World
☒	Left	CS2	[-0.265153 0.174114 1.05]	m	World	World
☒	Right	CS3	[-0.205153 -0.0308858 1.065]	m	World	World

OK Cancel Help Apply

(c)

Figure 3.27. (a) Orthesis Base Block Parameters (b) Orthesis_upperarm Block Parameters (c) Orthesis_forearm Block Parameters

Orthesis base body block is connected to rootpart body block from CS2 point with weld joint (Figure 3.26). This block is created by Simmechanics Link Software while exporting the design in Solidworks to Matlab. The parameters of this block are shown in Figure 3.28.

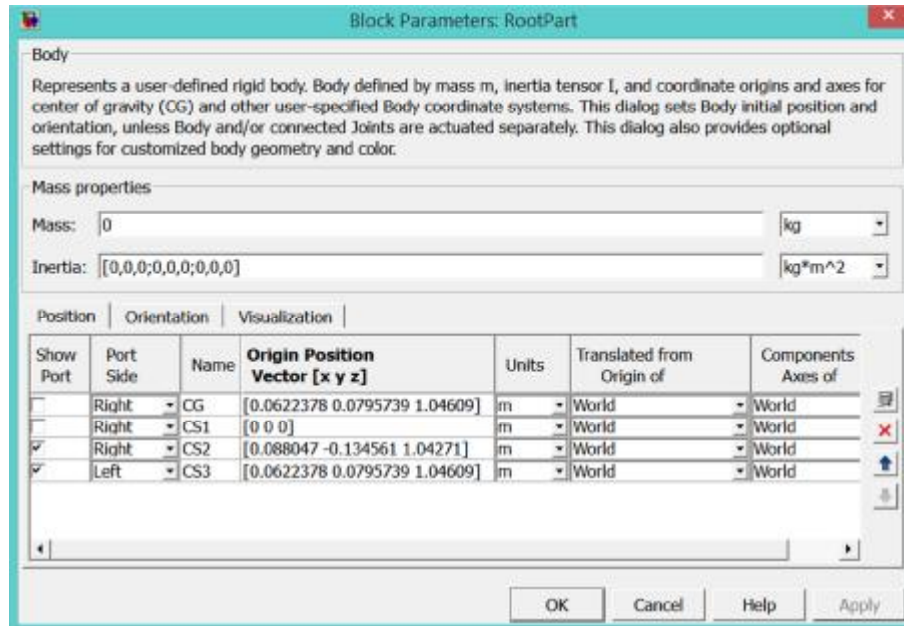


Figure 3.28. RootPart Block Parameters

No change was made on the block since the block did not have any mass and it did not have any effect on simulation as well.

RootPart was connected to RootGround with weld joint as shown in Figure 3.26. The parameters of RootGround block is shown in Figure 3.29.

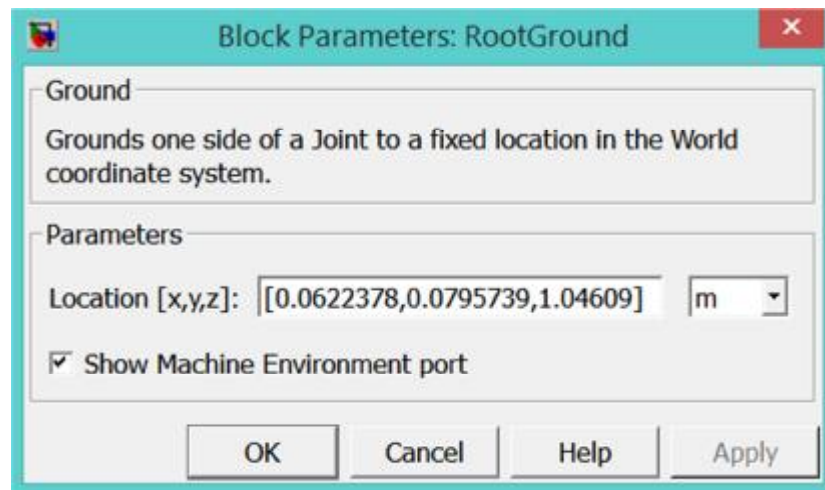


Figure 3.29. RootGround Block Parameters

At least one Ground block was connected to a Machine Environment for a valid SimMechanics. The block parameters of Machine Environment are shown in

Figure 3.30. Within Machine Environment Block gravity was defined in y direction as -9.81 m/s^2 . Forward Dynamics was selected as the analysis mode.

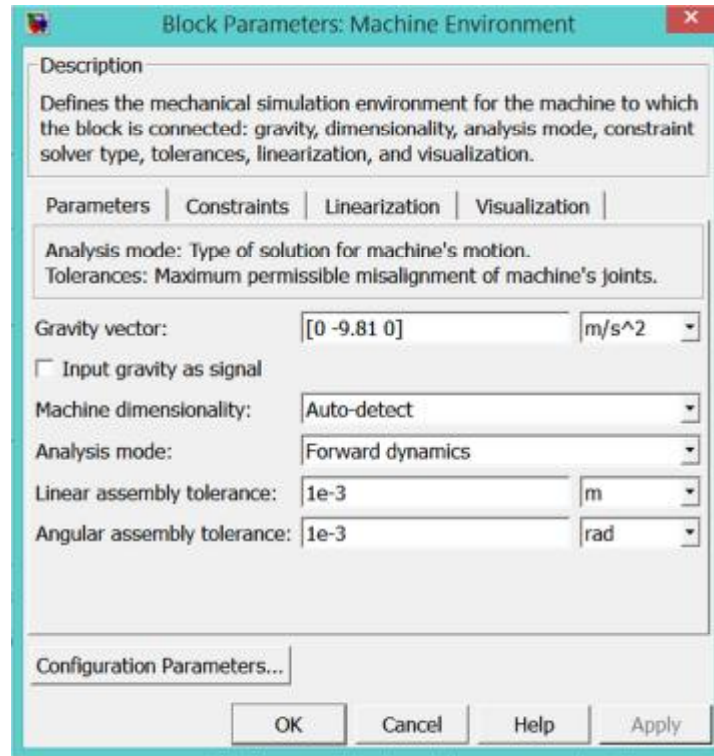


Figure 3.30. Machine Environment Block Parameters

3.2. Controller Design

3.2.1. Computed Torque Control

The control of the orthesis and HAMM in this thesis is made with the Computed-torque control method. Computed-Torque Control is a model-based control method which uses the dynamics model of the system to compute the control torque signals that are input of the system (Lewis et al., 2004). This controller generally performs well when the robot arm parameters are known fairly accurately. The reason why computed-torque control is preferred as a control method is that the mechanical arm parameters of both orthesis and HAMM are known accurately and this control method is convenient to be tested in simulation environment.

Designed controller consists of two parts named inner loop and outer loop as shown in Figure 3.31 where u is the Outer Loop control signal, τ is joint actuator torques, q and $\dot{q}$ are joint position and velocity and $\ddot{q}_d$ denotes the desired joint acceleration. Inner loop is Arm Dynamics, outer loop is PID controller.

Before controlling the HAMM and orthesis, it is necessary to know the desired trajectory for performing a task. There are many issues associated with the trajectory planning problem, such as avoiding obstacles and making sure that the planned trajectory does not require exceeding joint position limitations of the robots. To reduce the control problem to its basic components, in this controller structure, it was assumed that the one of the main control objective is to move the orthesis and HAMM along aprescribed desired trajectory (q_d).

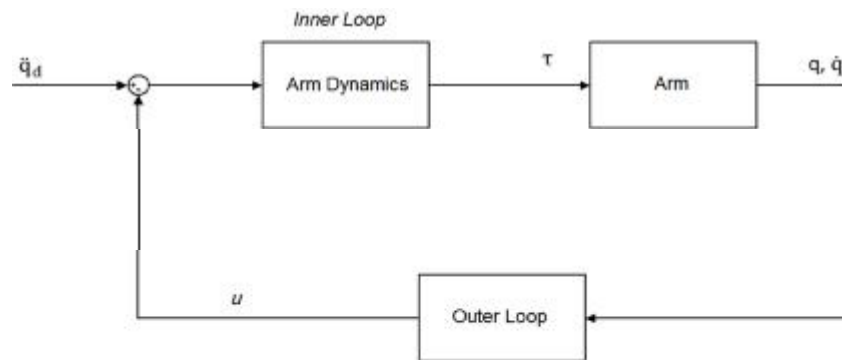


Figure 3.31. Computed Torque Control Block Diagram

3.2.1.1. The Inner Loop

The robot Arm Dynamics are (Lewis et al., 2004)

$$M(q)\ddot{q} + V(q, \dot{q}) + G(q) + \tau_{dis} = \tau \quad (3.1)$$

with q the joint position (in degrees or radians), $\dot{q}$ the first time derivative of the q (angular velocity), $\ddot{q}$ the second time derivative of q (angular acceleration), $M(q)$ the inertia matrix, $V(q, \dot{q})$ the Coriolis/centripetal vector, and $G(q)$ the gravity vector, τ the actuator control torque and τ_{dis} a disturbance.

The friction force of Arm Dynamics is ignored since this research is made in simulation environment. τ_{dis} (disturbance) is included into the arm dynamics as a 10 percent of the computed torque without disturbance effect.

3.2.1.2. The Outer PID Loop

General structure of PID controller output signal is

$$u(t) = -(K_p e + K_v \dot{e} + K_i \varepsilon(t)) \quad (3.2)$$

with e tracking error; $e = (q_d - q)$, $\dot{e} = \dot{q}_d - \dot{q}$ and $\varepsilon(t)$ the integral of the tracking error $e(t)$. K_p , K_v and K_i are proportional, derivative and integral gains of the PID controller respectively. These gain parameters were tuned manually to obtain the best performance results. PID gain parameters are given in the Table 3.3.

Table 3.3. PID Gain Parameters

K_p	K_v	K_i
550 - 650	50 - 70	150 - 250

With the PID controller, the overall robot arm input becomes

$$M(q)(\ddot{q}_d + K_p e + K_v \dot{e} + K_i \varepsilon(t)) + V(q, \dot{q}) + G(q) + \tau_{dis} = \tau \quad (3.3)$$

with $\ddot{q} = \ddot{q}_d - u$ (Craig 2005).

$M(q)$, $V(q, \dot{q})$ and $G(q)$ vector values of robot arm dynamic model must be computed to apply the computed-torque control.

3.2.1.3. Robot Dynamics

The robot dynamisc is derived using Lagrangian mechanics in this thesis. In this approach, kinetic and potential energy equation of manipulator is derived first, then manipulator dynamic model is derived using Langrange's equation of motion. Two different dynamic models are created and used in this thesis.

3.2.1.3.(1). Lagrange's Equations of Motion

Lagrange's equation of motion for a conservative system are given by (Lewis et al., 2004)

$$\frac{d}{dt} \frac{\partial L}{\partial \dot{q}} - \frac{\partial L}{\partial q} = \tau \quad (3.4)$$

where q is an n-vector of generalized coordinates q_i , τ is an n-vector of generalized forces τ_i , and the Lagrangian (L) is the difference between the kinetic (K) and potential (P) energies

$$L = K - P \quad (3.5)$$

In this usage, q will be the joint-variable vector, consisting of joint angles θ_i ; (in degrees or radians) and joint offsets d_i (in meters). Then τ is a vector that has components n_i of torque (newton-meters) corresponding to the i-th joint.

3.2.1.3.(2). Dynamic Model 1

In the model 1, center of mass of the links was taken into consideration while deriving dynamic model of the orthosis. This model is shown in Figure 3.32.

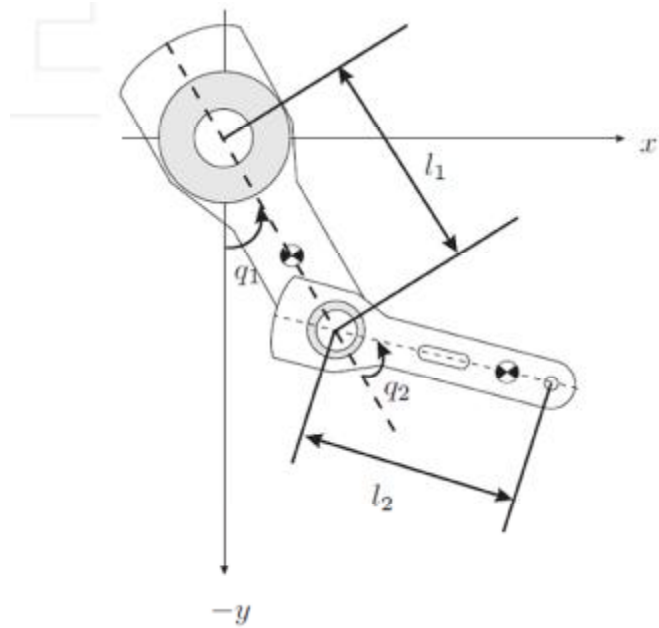


Figure 3.32. 2 Link Robot Arm for Model 1 (Sanchez-Sanchez et al. 2012)

For $i = 1, 2$, q_i denotes the joint angle, which also serves as a generalized coordinate; m_i denotes the mass of link i , l_i denotes the length of link i ; l_{ci} denotes the distance from the previous joint to the center of mass of link i ; and I_i denotes the moment of inertia of link i about an axis coming out of the page, passing through the center of mass of link i .

Potential and kinetic energy quantities which are necessary for Lagrange's equation are derived first to get the dynamic model.

The first step consists of obtaining the forward kinematics of the system described in Figure 3.32; the forward kinematics are (Sanchez-Sanchez et al. 2012):

$$v = \begin{bmatrix} l_1 \cos q_1 + l_{c2} \cos(q_1 + q_2) & l_{c2} \cos(q_1 + q_2) \\ l_1 \sin q_1 + l_{c2} \sin(q_1 + q_2) & l_{c2} \sin(q_1 + q_2) \end{bmatrix} \begin{bmatrix} \dot{q}_1 \\ \dot{q}_2 \end{bmatrix}$$

The square of velocity is represented for link1 and link 2 as;

$$v_1^2 = l_{c1}^2 + \dot{q}_1^2$$

$$v_2^2 = l_1^2 \dot{q}_1^2 + l_{c2}^2 (\dot{q}_1 + \dot{q}_2)^2 + 2l_1 l_{c2} \dot{q}_1 \cos q_2 (\dot{q}_1 + \dot{q}_2)$$

Using the definition of the kinetic energy; $K = \frac{1}{2}mv^2 + \frac{1}{2}I\dot{q}^2$ (Sanchez-Sanchez et al. 2012), kinetic energy K was calculated for Link 1 and Link 2 as

$$K_1 = \frac{1}{2}(m_1 l_{c1}^2 + I_1) \dot{q}_1^2$$

$$K_2 = \frac{1}{2}m_2 l_1^2 \dot{q}_1^2 + m_2 l_1 l_{c2} \dot{q}_1 \cos q_2 (\dot{q}_1 + \dot{q}_2) + \frac{1}{2}(m_2 l_{c2}^2 + I_2)(\dot{q}_1 + \dot{q}_2)^2$$

The total kinetic energy K is defined as

$$K = K_1 + K_2$$

Then the potential energy P was calculated for both links ;

$$P_1 = -m_1 l_{c1} g \cos q_1$$

$$P_2 = -m_2 l_1 g \cos q_1 - m_2 l_{c2} g \cos(q_1 + q_2).$$

The total potential energy P is obtained as

$$P = P_1 + P_2.$$

The next step is to calculate the Lagrangian L , which was defined as the difference between the total kinetic and total potential energy ;

$$L = K - P$$

$$L = m_2 l_1 l_{c2} \dot{q}_1 \cos q_2 (\dot{q}_1 + \dot{q}_2) + m_1 l_{c1} g \cos q_1 + m_2 l_1 g \cos q_1 + \frac{1}{2} (m_1 l_{c1}^2 + I_1 + m_2 l_1^2) \dot{q}_1^2 + \frac{1}{2} (m_2 l_{c2}^2 + I_2) (\dot{q}_1 + \dot{q}_2)^2 + m_2 l_{c2} g \cos (q_1 + q_2).$$

The terms needed for Equation (3.4) were calculated as

$$\begin{aligned} \frac{\partial L}{\partial \dot{q}_1} &= 2m_2 l_1 l_{c2} \dot{q}_1 \cos q_2 + m_2 l_1 l_{c2} \dot{q}_2 \cos q_2 + m_2 l_{c2}^2 \dot{q}_2 \\ &\quad + (m_1 l_{c1}^2 + I_1 + m_2 l_1^2 + m_2 l_{c2}^2) \dot{q}_1 + I_2 \dot{q}_1 + I_2 \dot{q}_2 \\ \frac{\partial L}{\partial \dot{q}_2} &= m_2 l_1 l_{c2} \dot{q}_1 \cos q_2 + m_2 l_{c2}^2 \dot{q}_1 + I_2 \dot{q}_1 + I_2 \dot{q}_2 \\ &\quad + m_2 l_{c2}^2 \dot{q}_2 \\ \frac{d}{dt} \frac{\partial L}{\partial \dot{q}_1} &= 2m_2 l_1 l_{c2} \ddot{q}_1 \cos q_2 - 2m_2 l_1 l_{c2} \dot{q}_2 \dot{q}_1 \sin q_2 + m_2 l_1 l_{c2} \ddot{q}_2 \cos q_2 \\ &\quad - m_2 l_1 l_{c2} \dot{q}_2^2 \sin q_2 + m_2 l_{c2}^2 \ddot{q}_2 + (m_1 l_{c1}^2 + I_1 + m_2 l_1^2 + m_2 l_{c2}^2) \ddot{q}_1 + \\ &\quad I_2 (\ddot{q}_1 + \ddot{q}_2) \\ \frac{d}{dt} \frac{\partial L}{\partial \dot{q}_2} &= m_2 l_1 l_{c2} \ddot{q}_1 \cos q_2 - m_2 l_1 l_{c2} \dot{q}_1 \dot{q}_2 \sin q_2 + m_2 l_{c2}^2 \ddot{q}_1 + I_2 \ddot{q}_1 + I_2 \ddot{q}_2 \\ &\quad + m_2 l_{c2}^2 \ddot{q}_2 \\ \frac{\partial L}{\partial q_1} &= -m_1 l_{c1} g \sin q_1 - m_2 l_1 g \sin q_1 - m_2 l_{c2} g \sin (q_1 + q_2) \\ \frac{\partial L}{\partial q_2} &= -m_2 l_1 l_{c2} \dot{q}_1^2 \sin q_2 - m_2 l_1 l_{c2} \dot{q}_1 \dot{q}_2 \sin q_2 - m_2 l_{c2} g \sin (q_1 + q_2) \end{aligned}$$

and

$$\begin{aligned} \tau_1 &= [2m_2 l_1 l_{c2} \cos q_2 + m_1 l_{c1}^2 + m_2 l_1^2 + m_2 l_{c2}^2] \ddot{q}_1 + [I_1 + I_2] \ddot{q}_1 \\ &\quad + [m_2 l_1 l_{c2} \cos q_2 + m_2 l_{c2}^2 + I_2] \ddot{q}_2 - 2m_2 l_1 l_{c2} \dot{q}_2 \dot{q}_1 \sin q_2 - m_2 l_1 l_{c2} \dot{q}_2^2 \sin q_2 \\ &\quad + [m_1 l_{c1} + m_2 l_1] g \sin q_1 + m_2 l_{c2} g \sin (q_1 + q_2) \\ \tau_2 &= [m_2 l_1 l_{c2} \cos q_2 + m_2 l_{c2}^2 + I_2] \ddot{q}_1 + [m_2 l_{c2}^2 + I_2] \ddot{q}_2 \\ &\quad + m_2 l_1 l_{c2} \dot{q}_1^2 \sin q_2 + m_2 l_{c2} g \sin (q_1 + q_2). \end{aligned}$$

The vectorial representation of τ_1 and τ_2 are defined as follows ;

$$M(q) \begin{bmatrix} \ddot{q}_1 \\ \ddot{q}_2 \end{bmatrix} + \begin{bmatrix} -2m_2l_1l_{c2}\dot{q}_2\sin q_2 & -m_2l_1l_{c2}\dot{q}_2\sin q_2 \\ m_2l_1l_{c2}\dot{q}_1\sin q_2 & 0 \end{bmatrix} \begin{bmatrix} \dot{q}_1 \\ \dot{q}_2 \end{bmatrix} + \begin{bmatrix} (m_1l_{c1} + m_2l_1)g\sin q_1 + m_2l_{c2}g\sin(q_1 + q_2) \\ m_2l_{c2}g\sin(q_1 + q_2) \end{bmatrix} = \begin{bmatrix} \tau_1 \\ \tau_2 \end{bmatrix}$$

where

$$M(q) = \begin{bmatrix} m_1l_{c1}^2 + m_2l_1^2 + m_2l_{c2}^2 + 2m_2l_1l_{c2}\cos q_2 + I_1 + I_2 & m_2l_{c2}^2 + m_2l_1l_{c2}\cos q_2 + I_2 \\ m_2l_{c2}^2 + m_2l_1l_{c2}\cos q_2 + I_2 & m_2l_{c2}^2 + I_2 \end{bmatrix}$$

$$V(q, \dot{q}) = \begin{bmatrix} -2m_2l_1l_{c2}\dot{q}_2\sin q_2 & -m_2l_1l_{c2}\dot{q}_2\sin q_2 \\ m_2l_1l_{c2}\dot{q}_1\sin q_2 & 0 \end{bmatrix}$$

$$G(q) = \begin{bmatrix} (m_1l_{c1} + m_2l_1)g\sin q_1 + m_2l_{c2}g\sin(q_1 + q_2) \\ m_2l_{c2}g\sin(q_1 + q_2) \end{bmatrix}.$$

These manipulator dynamics are in the standard form

$$M(q)\ddot{q} + V(q, \dot{q})\dot{q} + G(q) = \tau \quad (3.6)$$

with $M(q)$ the inertia matrix, $V(q, \dot{q})$ the Coriolis/centripetal vector, and $G(q)$ the gravity vector.

3.2.1.3.(3). Dynamic Model 2

Link masses are defined as point masses at the distal ends of the link in Model 2. This model is shown in Figure 3.33.

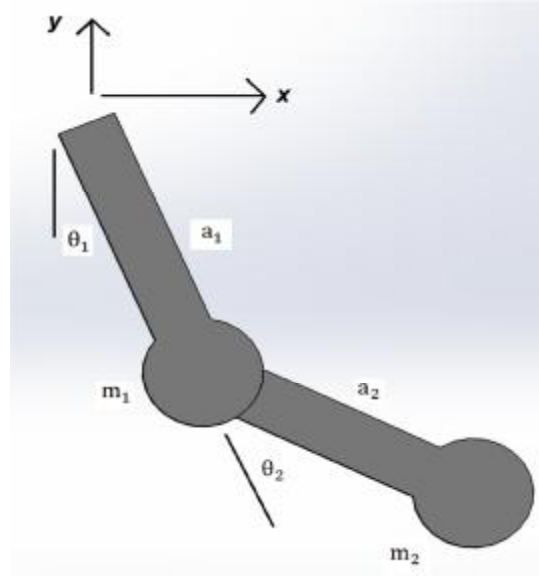


Figure 3.33. Two Link Robot Arm with Point Mass, Model 2

a_i represents link lengths, m_i represents point link masses, and θ_i represents joint angles. Because of the point mass assumption, inertia tensor written at the center of the mass for each link is zero matrix.

Joint variables for two link planar manipulator in Figure 3.33 are (Niku 2001)

$$q = [\theta_1 \quad \theta_2]^T$$

and the generalized force vector is

$$\tau = [\tau_1 \quad \tau_2]^T$$

with τ_1 , and τ_2 torques supplied by the actuators.

As in the Model 1, potential and kinetic energy quantities which are necessary for Lagrange's equation are calculated first to get the dynamic model.

For link 1 total kinetic and potential energies due to the angular motion $\dot{\theta}_1$ are

$$K_1 = \frac{1}{2} m_1 a_1^2 \dot{\theta}_1^2$$

$$P_1 = -m_1 g a_1 \cos \theta_1.$$

For link 2

$$x_2 = a_1 \sin \theta_1 + a_2 \sin(\theta_1 + \theta_2)$$

$$y_2 = -a_1 \cos \theta_1 - a_2 \cos(\theta_1 + \theta_2)$$

$$\dot{x}_2 = a_1 \dot{\theta}_1 \cos \theta_1 + a_2 (\dot{\theta}_1 + \dot{\theta}_2) \cos(\theta_1 + \theta_2)$$

$$\dot{y}_2 = a_1 \dot{\theta}_1 \sin \theta_1 + a_2 (\dot{\theta}_1 + \dot{\theta}_2) \sin(\theta_1 + \theta_2)$$

so that the velocity squared is

$$v_2^2 = \dot{x}_2^2 + \dot{y}_2^2 = a_1^2 \dot{\theta}_1^2 + a_2^2 (\dot{\theta}_1 + \dot{\theta}_2)^2 + 2a_1 a_2 (\dot{\theta}_1^2 + \dot{\theta}_1 \dot{\theta}_2) \cos \theta_2.$$

Therefore, the kinetic energy for link 2 is

$$K_2 = \frac{1}{2} m_2 v_2^2 = \frac{1}{2} m_2 a_1^2 \dot{\theta}_1^2 + \frac{1}{2} m_2 a_2^2 (\dot{\theta}_1 + \dot{\theta}_2)^2 + m_2 a_1 a_2 (\dot{\theta}_1^2 + \dot{\theta}_1 \dot{\theta}_2) \cos \theta_2.$$

The potential energy for link 2 is

$$P_2 = -m_2 g [a_1 \cos \theta_1 + a_2 \cos(\theta_1 + \theta_2)].$$

The Lagrangian for the entire arm is

$$\begin{aligned} L = K - P &= K_1 + K_2 - P_1 - P_2 \\ &= \frac{1}{2} (m_1 + m_2) a_1^2 \dot{\theta}_1^2 + \frac{1}{2} m_2 a_2^2 (\dot{\theta}_1 + \dot{\theta}_2)^2 + m_2 a_1 a_2 (\dot{\theta}_1^2 + \dot{\theta}_1 \dot{\theta}_2) \cos \theta_2 \\ &\quad + (m_1 + m_2) g a_1 \cos \theta_1 + m_2 g a_2 \cos(\theta_1 + \theta_2). \end{aligned}$$

The terms needed for Equation (3.4) are

$$\begin{aligned}
\frac{\partial L}{\partial \dot{\theta}_1} &= (m_1 + m_2)a_1^2\dot{\theta}_1 + m_2a_2^2(\dot{\theta}_1 + \dot{\theta}_2) + m_2a_1a_2(2\dot{\theta}_1 + \dot{\theta}_2)\cos\theta_2 \\
\frac{d}{dt}\frac{\partial L}{\partial \dot{\theta}_1} &= (m_1 + m_2)a_1^2\ddot{\theta}_1 + m_2a_2^2(\ddot{\theta}_1 + \ddot{\theta}_2) + m_2a_1a_2(2\ddot{\theta}_1 + \ddot{\theta}_2)\cos\theta_2 \\
&\quad - m_2a_1a_2(2\dot{\theta}_1\dot{\theta}_2 + \dot{\theta}_2^2)\sin\theta_2 \\
\frac{\partial L}{\partial \theta_1} &= -(m_1 + m_2)ga_1\sin\theta_1 - m_2ga_2\sin(\theta_1 + \theta_2) \\
\frac{\partial L}{\partial \dot{\theta}_2} &= m_2a_2^2(\dot{\theta}_1 + \dot{\theta}_2) + m_2a_1a_2\dot{\theta}_1\cos\theta_2 \\
\frac{d}{dt}\frac{\partial L}{\partial \dot{\theta}_2} &= m_2a_2^2(\ddot{\theta}_1 + \ddot{\theta}_2) + m_2a_1a_2\ddot{\theta}_1\cos\theta_2 - m_2a_1a_2\dot{\theta}_1\dot{\theta}_2\sin\theta_2 \\
\frac{\partial L}{\partial \theta_2} &= -m_2a_1a_2(\dot{\theta}_1^2 + \dot{\theta}_1\dot{\theta}_2)\sin\theta_2 - m_2ga_2\sin(\theta_1 + \theta_2)
\end{aligned}$$

Finally, according to Lagrange's equation, the arm dynamics are given by the two coupled nonlinear differential equations

$$\begin{aligned}
\tau_1 &= [(m_1 + m_2)a_1^2 + m_2a_2^2 + 2m_2a_1a_2\cos\theta_2]\ddot{\theta}_1 + [m_2a_2^2 + m_2a_1a_2\cos\theta_2]\ddot{\theta}_2 \\
&\quad - m_2a_1a_2(2\dot{\theta}_1\dot{\theta}_2 + \dot{\theta}_2^2)\sin\theta_2 + (m_1 + m_2)ga_1\sin\theta_1 + m_2ga_2\sin(\theta_1 + \theta_2) \\
\tau_2 &= [m_2a_2^2 + m_2a_1a_2\cos\theta_2]\ddot{\theta}_1 + m_2a_2^2\ddot{\theta}_2 + m_2a_1a_2\dot{\theta}_1^2\sin\theta_2 + m_2ga_2\sin(\theta_1 + \theta_2).
\end{aligned}$$

Writing the arm dynamics in vector form yields

$$\begin{aligned}
M(q) \begin{bmatrix} \ddot{\theta}_1 \\ \ddot{\theta}_2 \end{bmatrix} + \begin{bmatrix} -m_2a_1a_2(2\dot{\theta}_1\dot{\theta}_2 + \dot{\theta}_2^2)\sin\theta_2 \\ m_2a_1a_2\dot{\theta}_1^2\sin\theta_2 \end{bmatrix} \\
+ \begin{bmatrix} (m_1 + m_2)ga_1\sin\theta_1 + m_2ga_2\sin(\theta_1 + \theta_2) \\ m_2ga_2\sin(\theta_1 + \theta_2) \end{bmatrix} = \begin{bmatrix} \tau_1 \\ \tau_2 \end{bmatrix}
\end{aligned}$$

where

$$M(q) = \begin{bmatrix} (m_1 + m_2)a_1^2 + m_2a_2^2 + 2m_2a_1a_2\cos\theta_2 & m_2a_2^2 + m_2a_1a_2\cos\theta_2 \\ m_2a_2^2 + m_2a_1a_2\cos\theta_2 & m_2a_2^2 \end{bmatrix}$$

$$V(q, \dot{q}) = \begin{bmatrix} -m_2a_1a_2(2\dot{\theta}_1\dot{\theta}_2 + \dot{\theta}_2^2)\sin\theta_2 \\ m_2a_1a_2\dot{\theta}_1^2\sin\theta_2 \end{bmatrix}$$

$$G(q) = \begin{bmatrix} (m_1 + m_2)ga_1\sin\theta_1 + m_2ga_2\sin(\theta_1 + \theta_2) \\ m_2ga_2\sin(\theta_1 + \theta_2) \end{bmatrix}$$

These manipulator dynamics are in the standard form

$$M(q)\ddot{q} + V(q, \dot{q}) + G(q) = \tau \quad (3.7)$$

with $M(q)$ the inertia matrix, $V(q, \dot{q})$ the Coriolis/centripetal vector, and $G(q)$ the gravity vector.

When Equation (3.6) and (3.7) are analysed, it is seen that two different manipulator dynamic equations are obtained for two different dynamic models.

For the 1st Dynamic Model;

$$M(q)(\ddot{q}_d + K_p e + K_v \dot{e} + K_i \varepsilon(t)) + V(q, \dot{q})\dot{q} + G(q) + \tau_{dis} = \tau \quad (3.8)$$

For the 2nd Dynamic Model;

$$M(q)(\ddot{q}_d + K_p e + K_v \dot{e} + K_i \varepsilon(t)) + V(q, \dot{q}) + G(q) + \tau_{dis} = \tau \quad (3.9)$$

PID Computed-Torque Control equations are obtained. These equations are used for the control of orthosis and HAMM.

3.2.2. Implementation of the Controller

PID Computed-torque control block diagram is shown in Figure 3.34 where $N(q, \dot{q}) = V(q, \dot{q}) + G(q)$

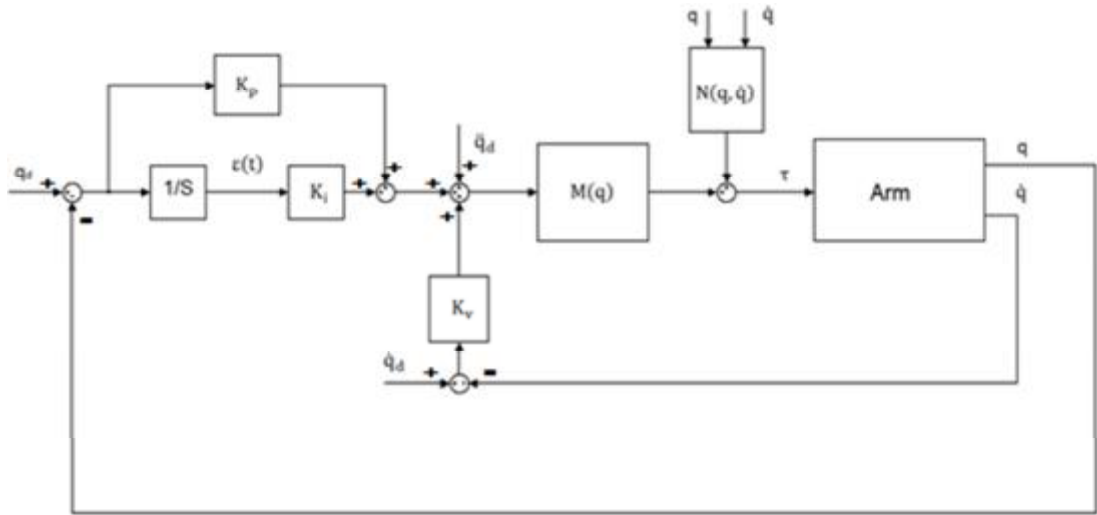


Figure 3.34. PID Computed-torque Control Block Diagram

Position, angular velocity and angular acceleration data of the joints of the HAMM or orthesis which controller will be used is needed to implement the Computed-torque Control. Angular acceleration data of the joints are obtained from the joints in the orthesis or HAMM mechanical models designed in Matlab using Simmechanics First Generation Joint Sensor Blocks. Parameter of Joint Sensor Block is shown in Figure 3.35.

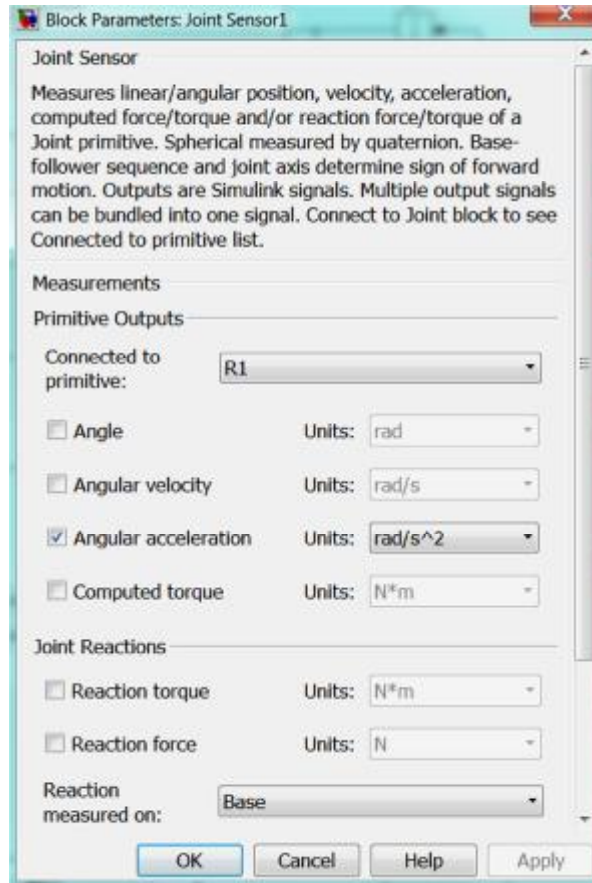


Figure 3.35. Joint Sensor Block Parameters

Angular acceleration is calculated as rad/s^2 . Joint angular velocity data is obtained from getting integral of angular acceleration data which are measured with sensors by using Matlab Integrator Function Block. Integrator Function Block parameters are shown in Figure 3.36.

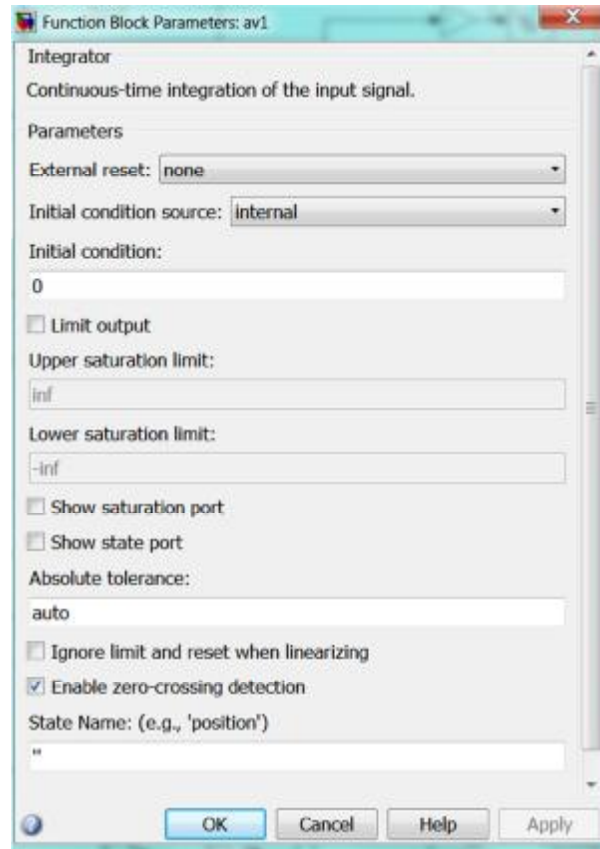


Figure 3.36. Integrator Function Block Parameters

Joint positions are obtained from taking the integral of the joint angular velocity data by using the same block. This position and angular velocity data are used as actual position (ap) and actual velocity (av) input of manipulator dynamic model.

Another data needed by Computed-torque control is the data of position, angular velocity and angular acceleration of desired trajectory. Desired trajectories are specified as sinus waveform by using Matlab Source Blocks. Desired angular velocity and desired angular acceleration values are 1st and 2nd derivative of desired position values. While finding the values of velocity and acceleration by using Matlab derivative function block, some problems are encountered in case of discontinuity in position signals. This situation causes a sudden change in the velocity signal and an infinite acceleration⁴. Infinite accelerations are not realistic

⁴<http://blogs.mathworks.com/seth/2012/02/01/applying-motion-to-simmechanics-models/>

and cause derivative errors in the simulation solver. To overcome this problem, desired acceleration and desired velocity data are obtained using Simulink Transfer Function as filtered derivative instead of derivation as shown in Figure 3.37.

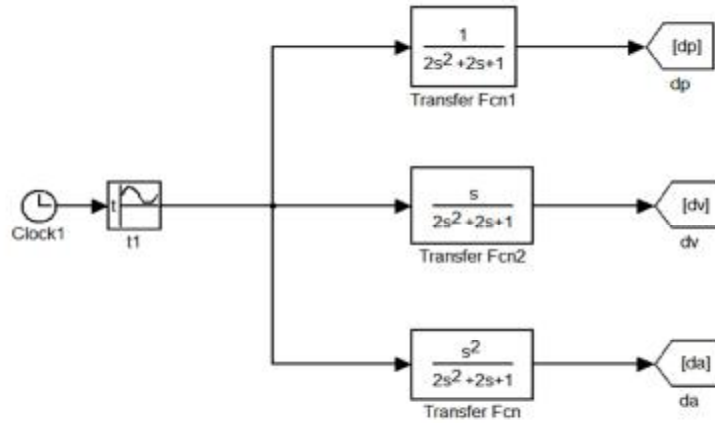


Figure 3.37. Filtered Derivative Operation

This obtained position, angular velocity and angular acceleration datas are used as desired position (dp) and desired velocity (dv) and desired acceleration (da) input of manipulator dynamic model. The integral of the tracking error $e(t)$ term ($\varepsilon(t)$) is calculated out of dynamic model as shown in Figure 3.39 and it is added as E1 and E2 terms to the dynamic model.

Equations of (3.8) and (3.9) in the previous section are created by using Matlab Function Blocks (Figure 3.38).

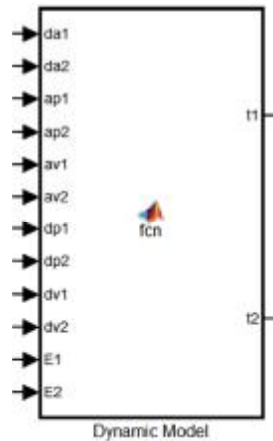


Figure 3.38. Matlab Function Block

In controller dynamic model 1 and dynamic model 2 equations which were written for orthosis and HAMM are given in Appendix.

PID Computed-torque control Matlab implementation for Orthosis and HAMM are shown in Figure 3.39 and Figure 3.40.

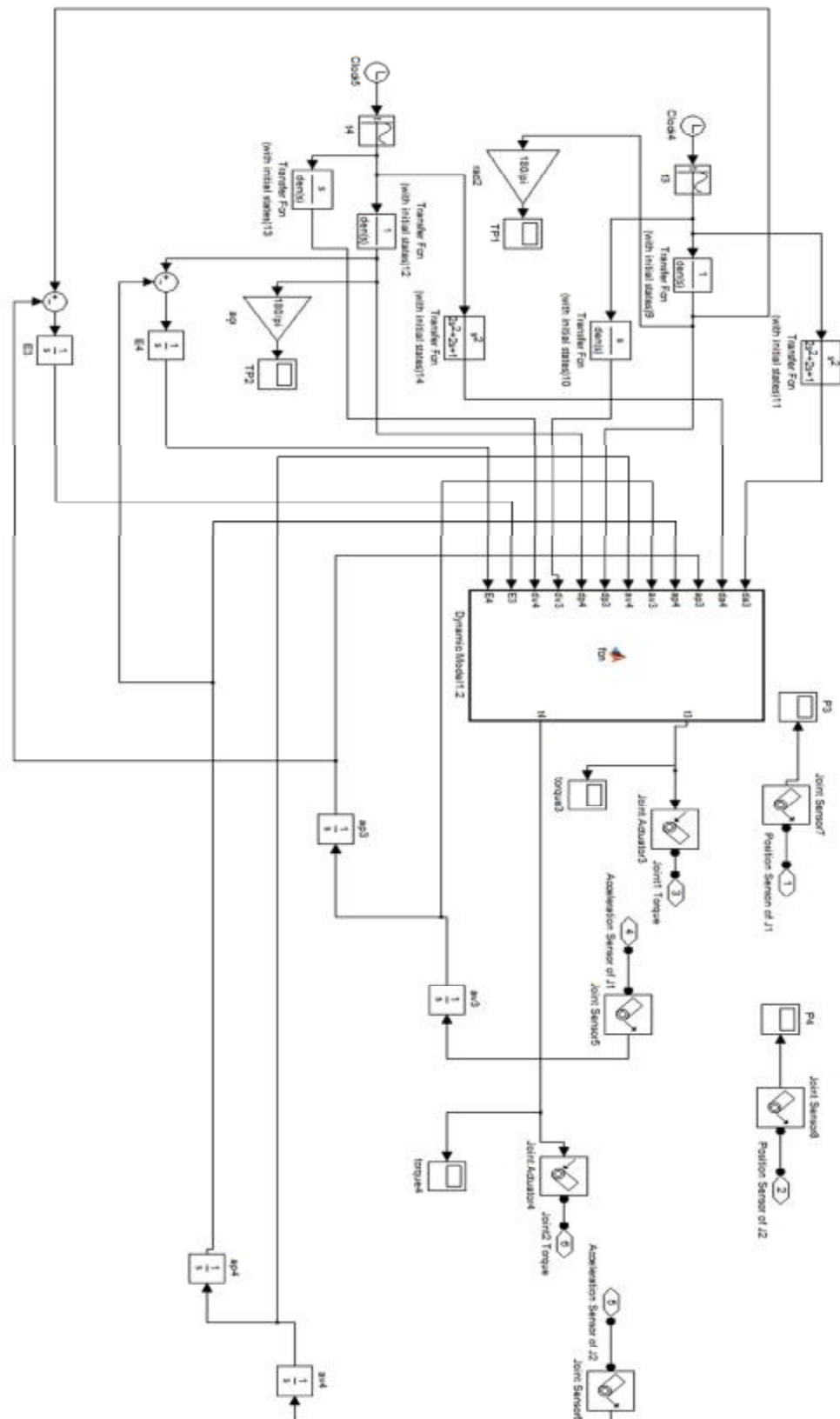


Figure 3.40. PID Computed-Torque Control Matlab Implementation For HAMM

3.2.3. Matlab Graphical User Interface (GUI)

Orthesis Resistive and Assistive rehabilitation modes and mode transitions are done with Graphical User Interface designed in Matlab. The designed GUI is shown in Figure 3.41.

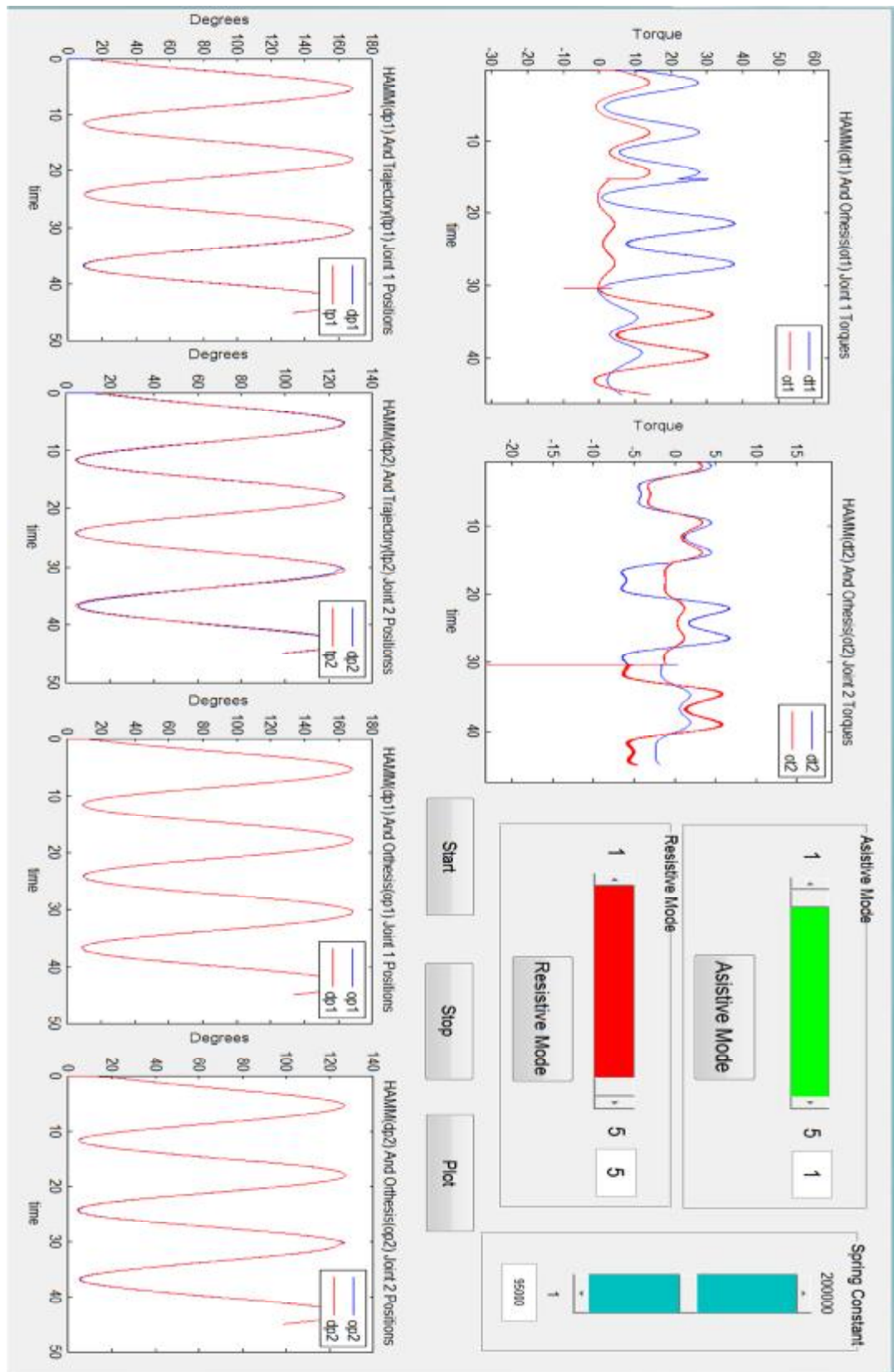


Figure 3.41. Matlab Graphical User Interface (GUI)

The designed GUI has 4 main sections. First section consists of 2 graphics (Figure 3.42).

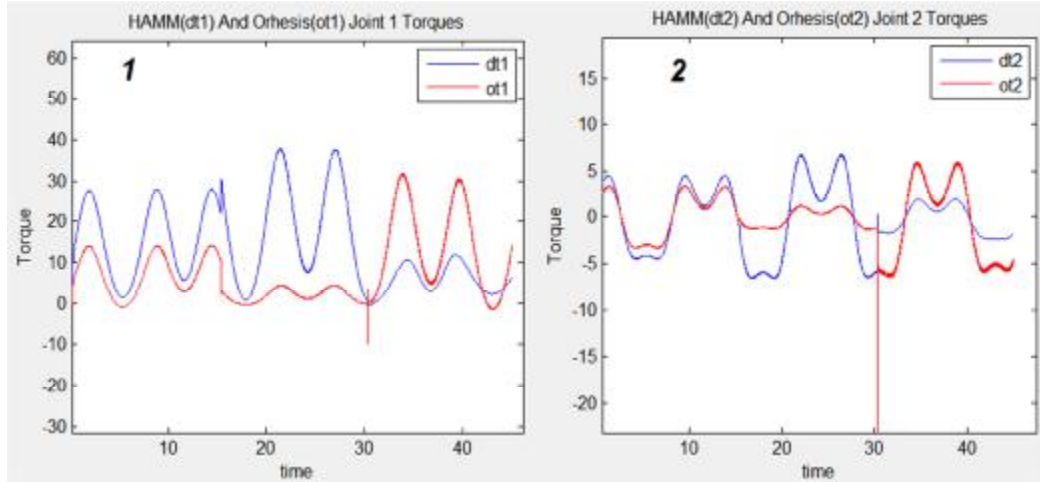


Figure 3.42. First Section of GUI

1st graphic shows first joint actuator torques of orthesis and HAMM. 2nd graphic shows second joint actuator torques of orthesis and HAMM. It is aimed to compare and analyse the joint torque of HAMM and orthesis during rehabilitation modes and mode transitions by showing orthesis and HAMM joint torques in the same graphic.

Second and third sections also consist of 2 graphics. The graphics of section 2 show joint positions of HAMM and desired trajectories of these joints (Figure 3.43).

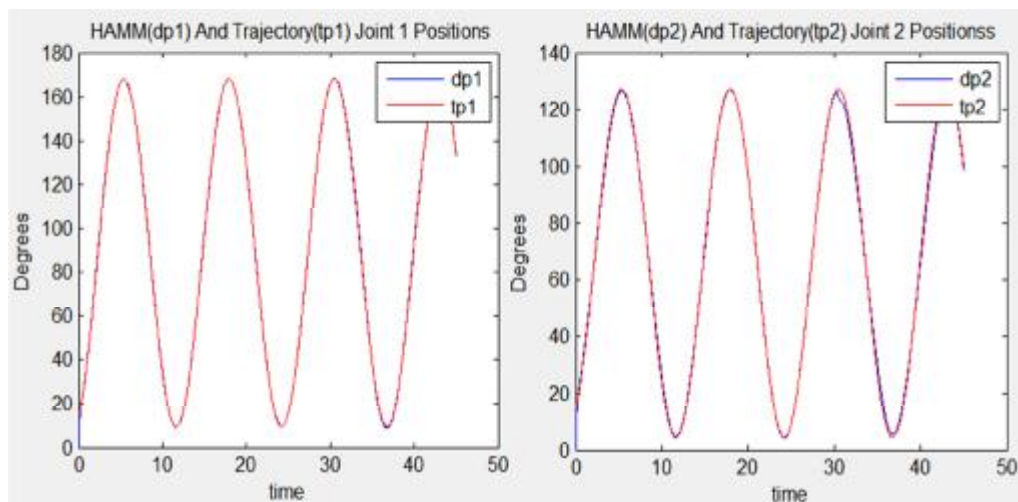


Figure 3.43. Second Section of GUI

The graphics of section 3 show the first and second joint positions of orthesis and HAMM (Figure 3.44).

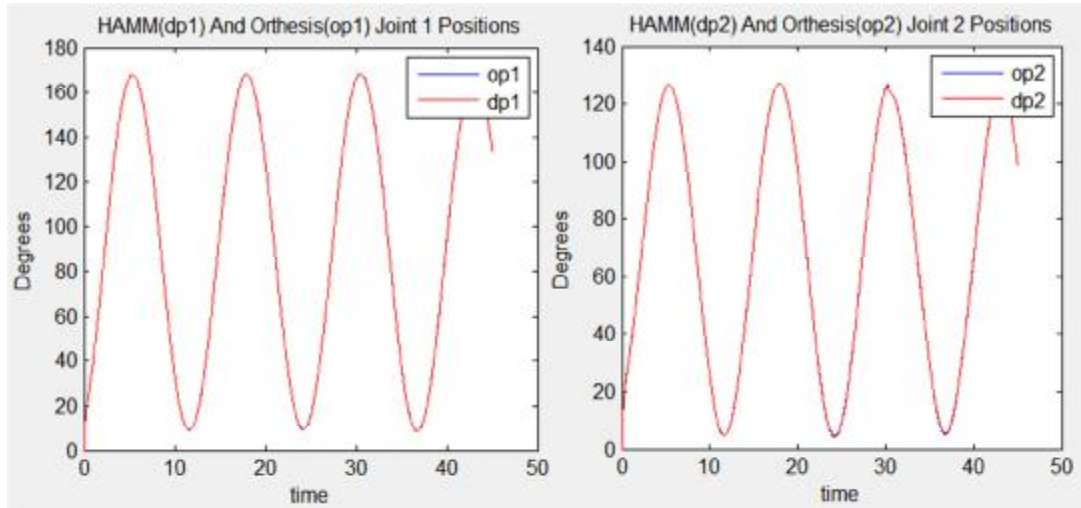


Figure 3.44. Third Section of GUI

4th section of the GUI is the simulation control panel which includes 3 push buttons named Start, Stop and Plot, resistive/assistive mode control panels and the slider which is used to adjust the spring constant of Body Spring & Damper block. Tightness of connection between orthesis and HAMM is changed by changing spring constant and the effect of this change to the results are observed.



Figure 3.45. 4th Section of GUI

Start push button was used to start the simulation and stop push button was used to stop the simulation. Plot push button draws the graphics stated in the first three sections to Axes of GUI.

Resistive and assistive modes are created by adding Simulink Gain Blocks to torque outputs of the orthosis PID computed torque control. Gain is 1 at the beginning of the simulation. Orthosis is at neither assistive nor resistive mode at this moment. Gain is adjusted larger than 1 as assistive mode is selected. It is aimed to make the orthosis assist HAMM to move and let the HAMM achieve the same movement with less torque in this way. The greater the Gain value, the more assist level of the orthosis is. In resistive mode, a negative impact is created on HAMM's movement by preferring the Gain value less than 1. It is aimed to make HAMM spend more torque to make the movement in this way. The less the Gain value, the more the resist level of the orthosis is.

Assistive mode control panel consists of toggle button and slider. Orthosis operates in assistive mode as the assistive mode toggle button is selected. To make the orthosis exit assistive mode and switch to resistive mode, resistive mode toggle button must be pressed. Slider selects the mode levels of assistive mode. Assistive mode consists of 5 levels. Level 1 is the level with the least assist, level 5 is the level with the highest assist. Gain values are selected as 2, 4, 6, 8, and 10 in order for each level from 1 to 5 of each mode.

Resistive mode control panel is like assistive mode control panel consists of toggle button and slider. Orthosis operates in resistive mode as the resistive mode toggle button is selected. To make the orthosis exit resistive mode and switch to assistive mode, assistive mode toggle button must be pressed. Slider selects the mode levels of resistive mode. Resistive mode consists of 5 levels. Level 5 is the level with the least challenging effect, level 1 is the level with the most challenging effect. Gain values are selected as 0.5, 0.4, 0.3, 0.2, and 0.1 in order for each level from 5 to 1 of each mode.

Three dimensional view of the simulation which was performed by using GUI is shown in Figure 3.46.

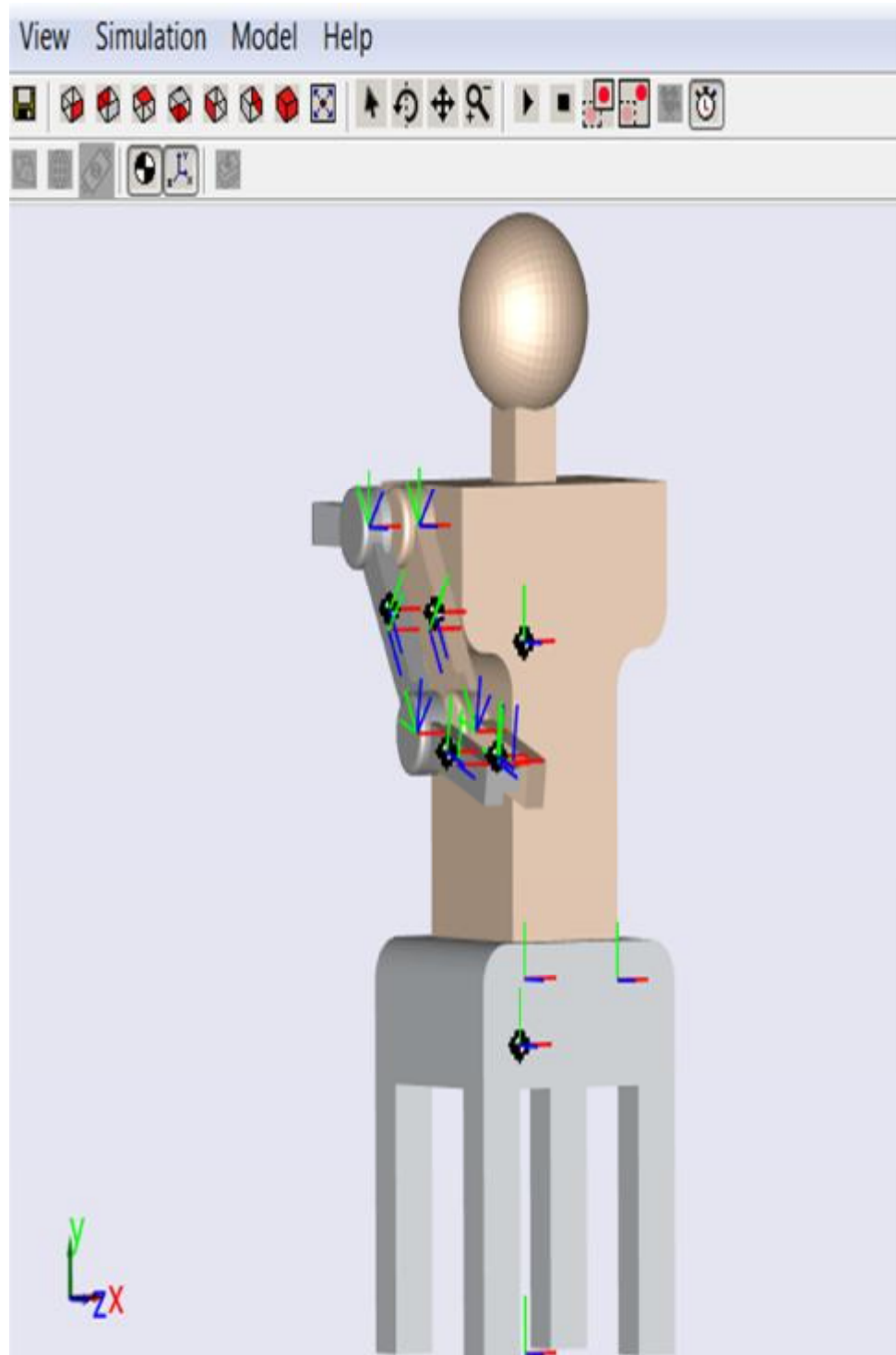


Figure 3.46. 3D View of the Simulation

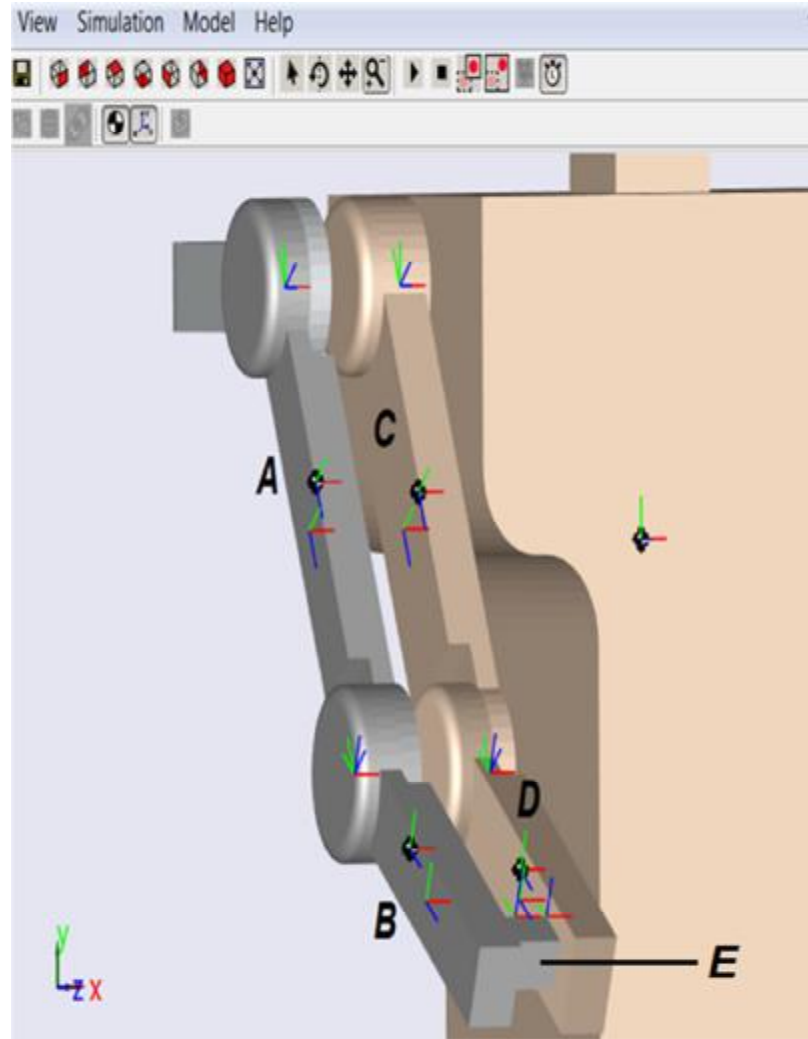


Figure 3.47. Moving Parts of HAMM and Orthosis

Part A which is seen in Figure 3.47 represents the Orthosis upperarm, part B represents the orthosis forearm, part C represents the HAMM upperarm, part D represents the HAMM forearm part and E represents the Coupling Block (Body Spring & Damper Block).

4. RESULTS AND DISCUSSION

4 different Simulink models were created by using two different dynamic models under the control of orthosis and HAM

M. Simulink models and orthosis and HAMM controllers used in these models are shown in Table 4.1. HAMM1 represents that dynamic model 1 is used in HAMM controller, HAMM2 represents that dynamic model 2 is used in HAMM controller, Orthosis1 means that dynamic model 1 is used in orthosis controller, Orthosis2 means that dynamic model 2 is used in orthosis controller. The results of these Simulink models will be given in this chapter.

Table 4.1. Simulink Models

Simulink Models	HAMM Controller	Orthosis Controller
Model 1	HAMM1	Orthosis1
Model 2	HAMM2	Orthosis2
Model 3	HAMM2	Orthosis1
Model 4	HAMM1	Orthosis2

Designed controller has assistive and resistive rehabilitation modes. Each mode has 5 levels. In assistive mode, level 1 represents the least, level 5 represents the highest assist. In resistive mode, level 5 represents the least challenging effect, level 1 represents the most challenging effect. Mode transitions during simulation are made with GUI as described in the previous chapter.

Same desired trajectories were applied to the orthosis and HAMM joints during simulations. Assistive and resistive modes of orthosis and performance of mode transitions for these trajectories were analysed. Two different desired trajectories were used for simulations. First desired trajectory is shown for two joints in Figure 4.1 and 4.2.

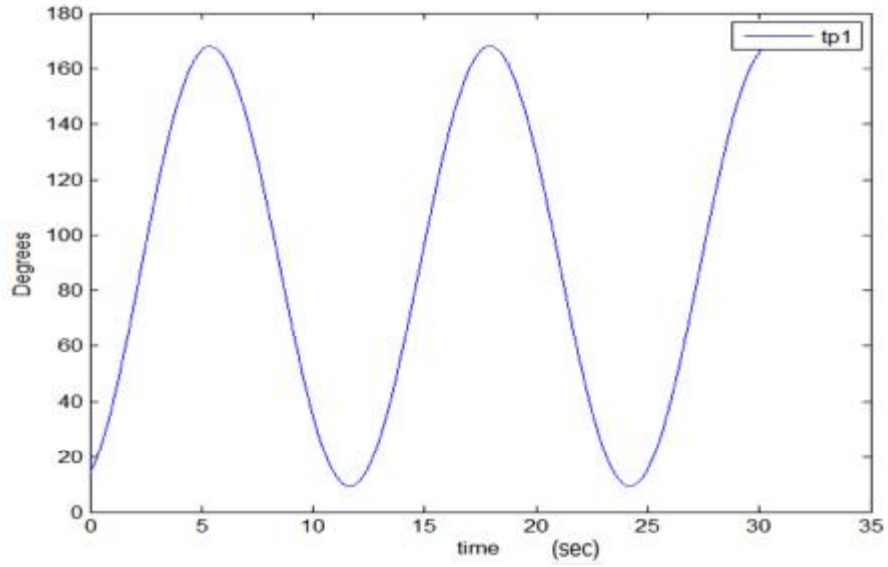


Figure 4.1. Desired trajectory (tp1) for the first joint

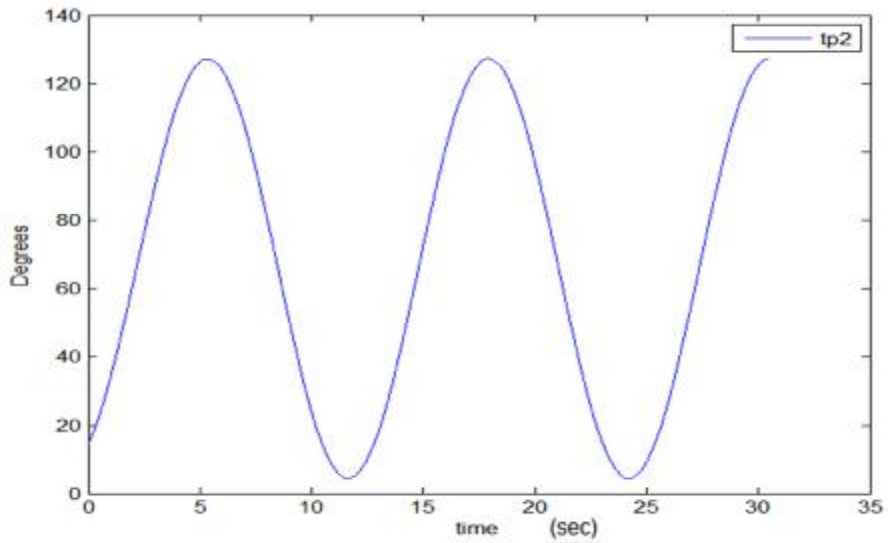


Figure 4.2. Desired trajectory (tp2) for the second joint

Gain constant value was set as 1 for orthosis joint actuators at the beginning of all simulations. Orthosis is in neither assistive nor resistive mode. This situation is stated as normal mode in results section.

4.1. Simulation Results of Model 1

The control of orthosis and HAMM was performed using dynamic model 1 (DM1) in 1st model. The torque values of HAMM joint 1 and joint 2 of the simulation using 1st model are given in Figures 4.3 and 4.4. The orthosis was operated in normal mode for first 10 seconds during simulation. The orthosis had been at assistive mode 4th level since t_1 time instant. The orthosis switched to resistive mode 1st level after t_2 time instant. t_1 and t_2 time instants are given in Figures below.

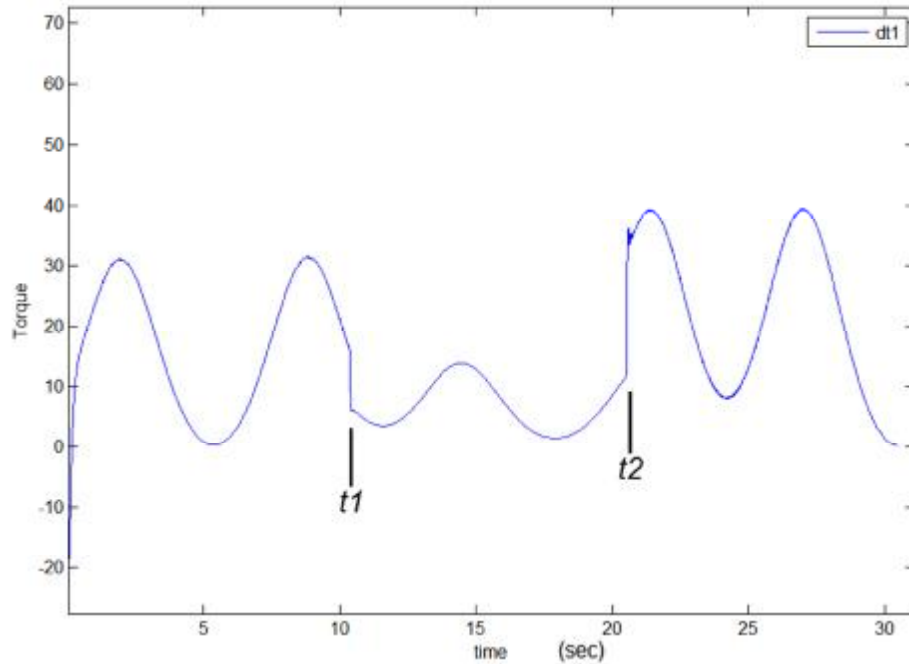


Figure 4.3. Model 1 HAMM Joint 1 Actuator Torques

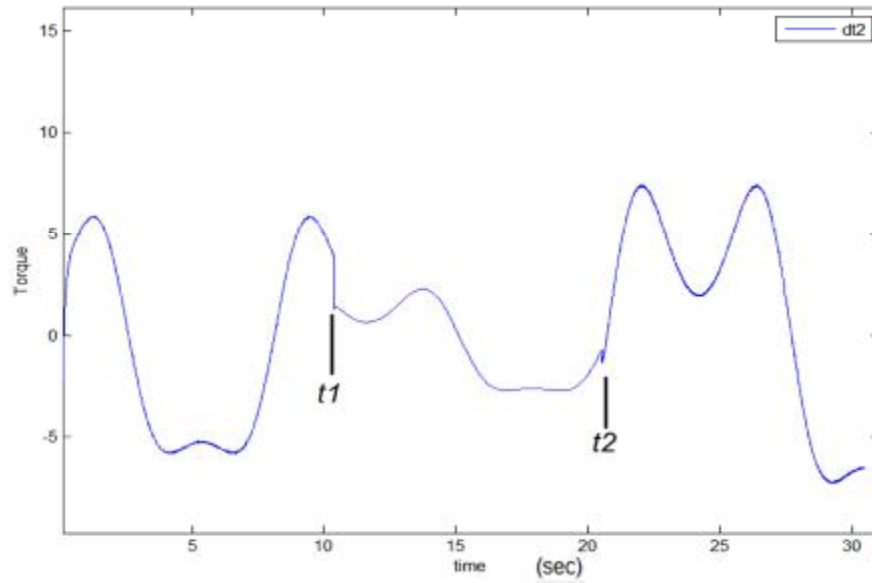


Figure 4.4. Model 1 HAMM Joint 2 Actuator Torques

Orthesis joint 1 and joint 2 actuator torques obtained from the same simulation are shown in Figures 4.5 and 4.6.

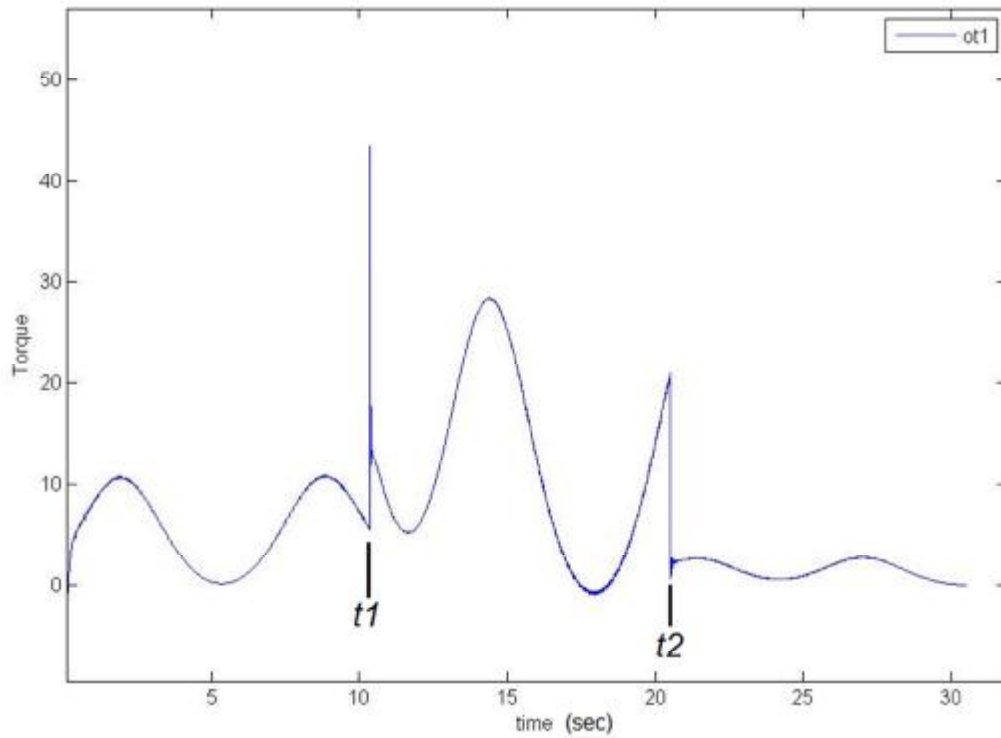


Figure 4.5. Model 1 Orthesis Joint 1 Actuator Torques

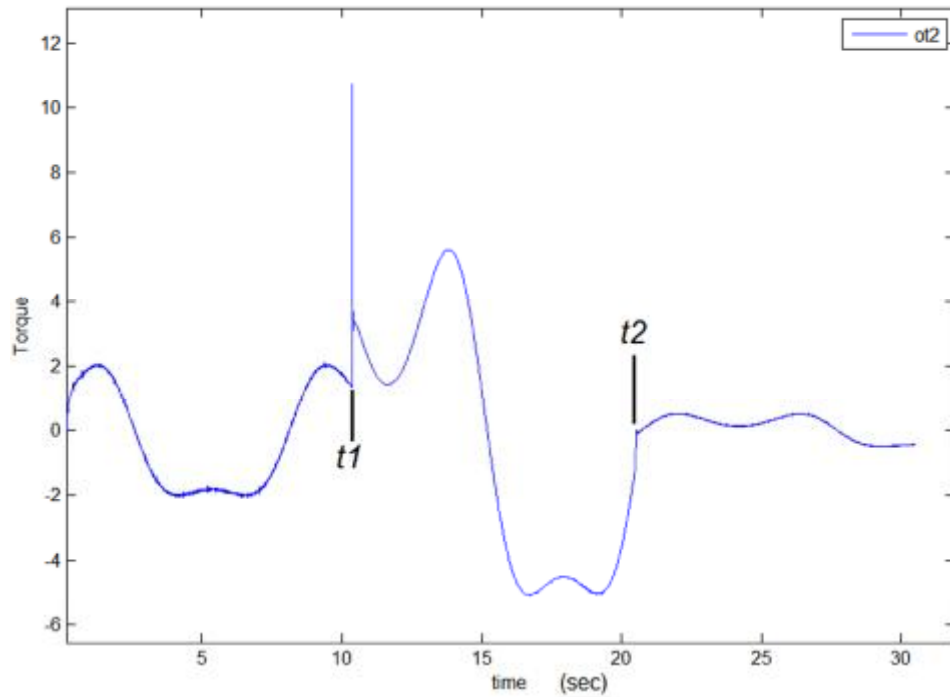
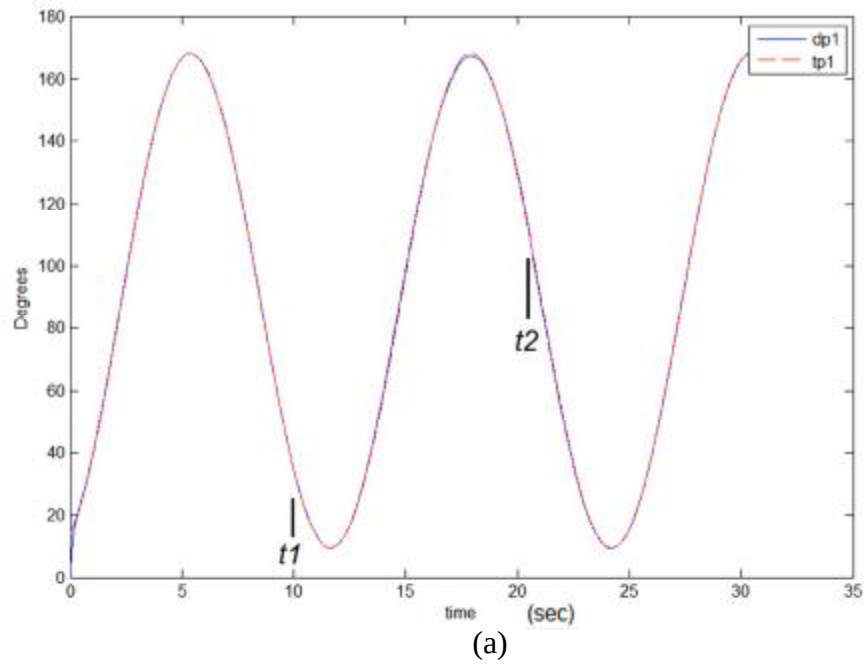


Figure 4.6. Model 1 Orthosis Joint 2 Actuator Torques

The desired trajectories for both joints and HAMM joint positions in simulation 1 are shown in Figure 4.7.



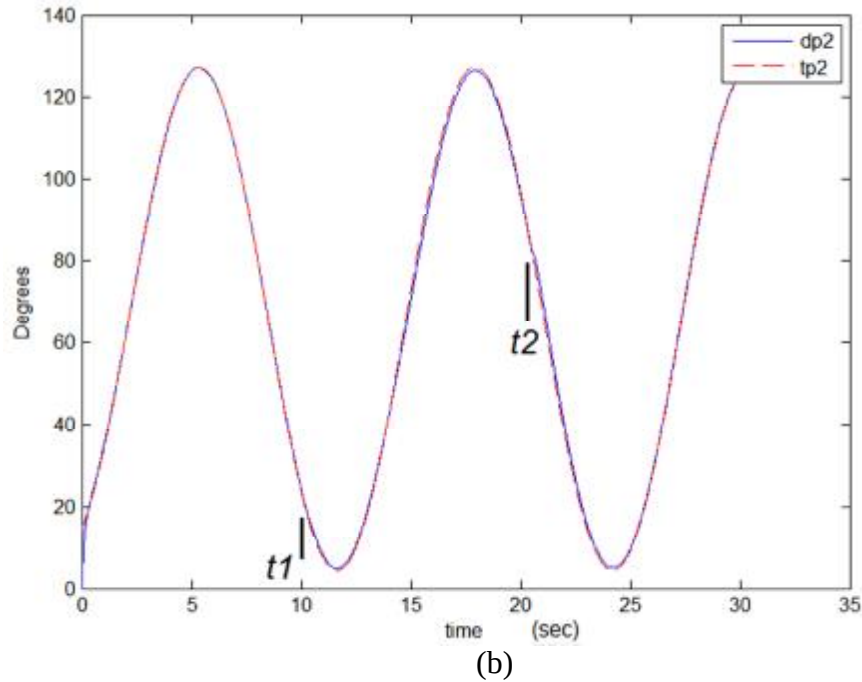


Figure 4.7. Model 1 (a) HAMMJoint1 (dp1) Position And Desired Trajectory (tp1) For Joint 1 (b) HAMM Joint 2 (dp2) Position And Desired Trajectory (tp2) For Joint 2

Average Error (AE) for HAMM and orthesis joint positions in simulation 1 are given in Figures 4.8 and 4.9.

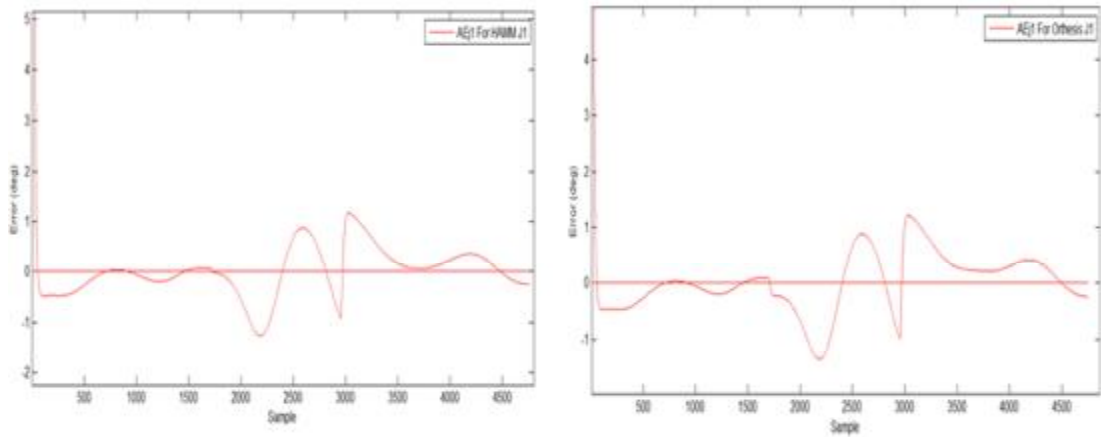


Figure 4.8. AE for HAMM and Orthesis Joint1 Position

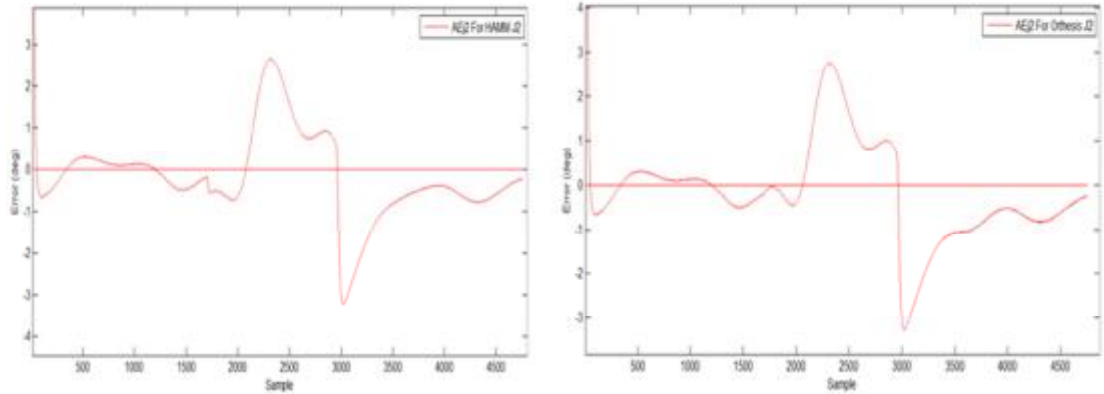
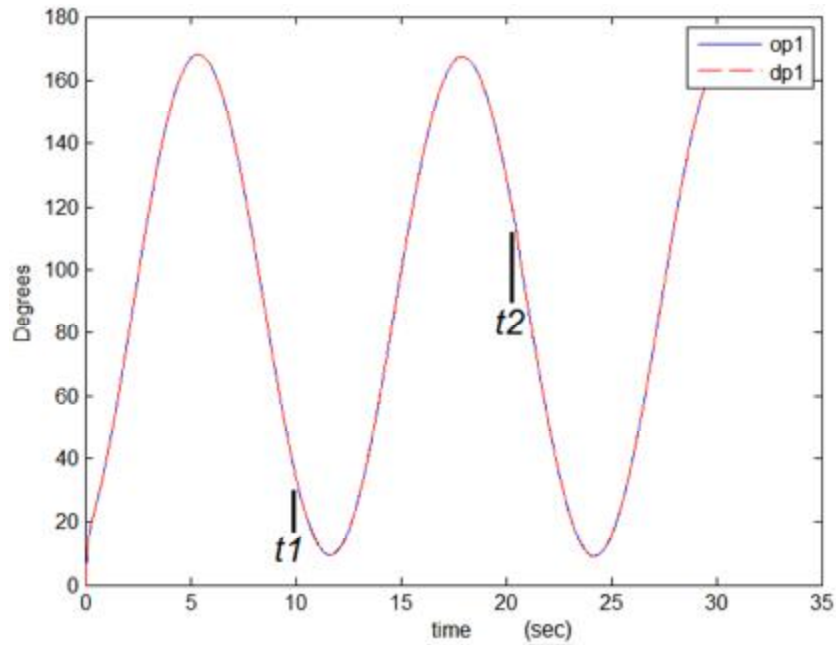


Figure 4.9. AE for HAMM and Orthesis Joint2 Position

AE values are very similar to each other for orthesis and HAMM joint positions as shown in Figure 4.8 and 4.9. So, only AE for HAMM joint positions values are given in the following results.

HAMM and orthesis joint positions obtained from simulation are shown in Figure 4.10.



(a)

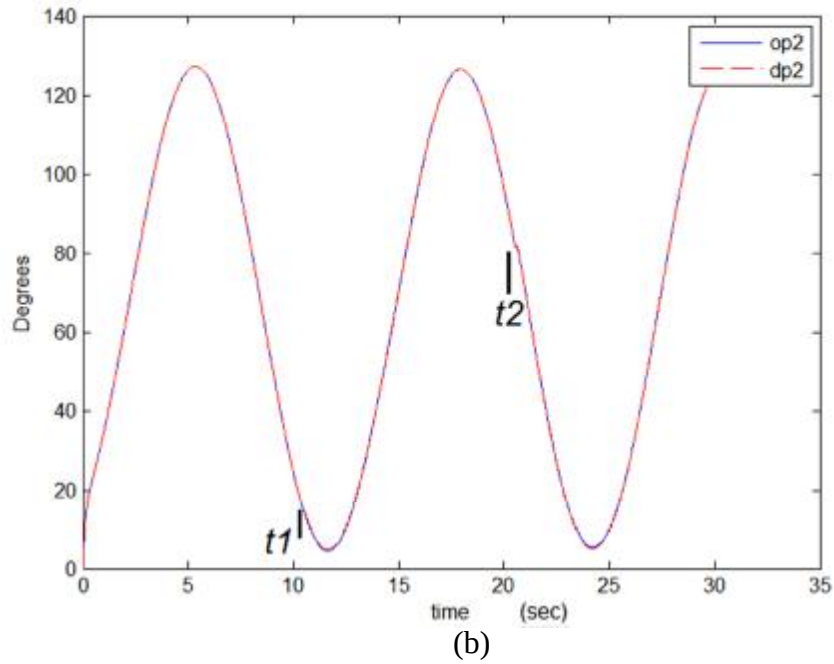


Figure 4.10. Model 1 (a) HAMM And Orthesis Joint1 (dp1, op1) Positions (b) HAMM And Orthesis Joint2 (dp2, op2) Positions

Analysing the results obtained from 1st model simulation, HAMM joint torques decrease at assistive mode. It can be understood from this situation that orthesis helped HAMM motion in assistive mode. That HAMM joint torques increase in resistive mode means orthesis had a challenging effect on HAMM's movement. It can be understood from Figure 4.10 that HAMM and orthesis moved along with each other during simulation. In Figure 4.7, it can be seen that as soon as the orthesis switched to resistive mode at t_2 , an error occurred at the HAMM 2nd position.

4.2. Simulation Results of Model 2

The control of orthesis and HAMM were made by using dynamic model 2 (DM2) in 2nd model. The HAMM joint 1 and joint 2 actuator torque values of the simulation which made by using 2nd model are given in Figures 4.11 and 4.12. The orthesis was operated in normal mode for first 10 seconds during simulation. The orthesis had been at resistive mode 3rd level starting from t_1 time instant. The

orthosis switched to assistive mode 5th level after t_2 time instant. t_1 and t_2 time instants are shown in Figures.

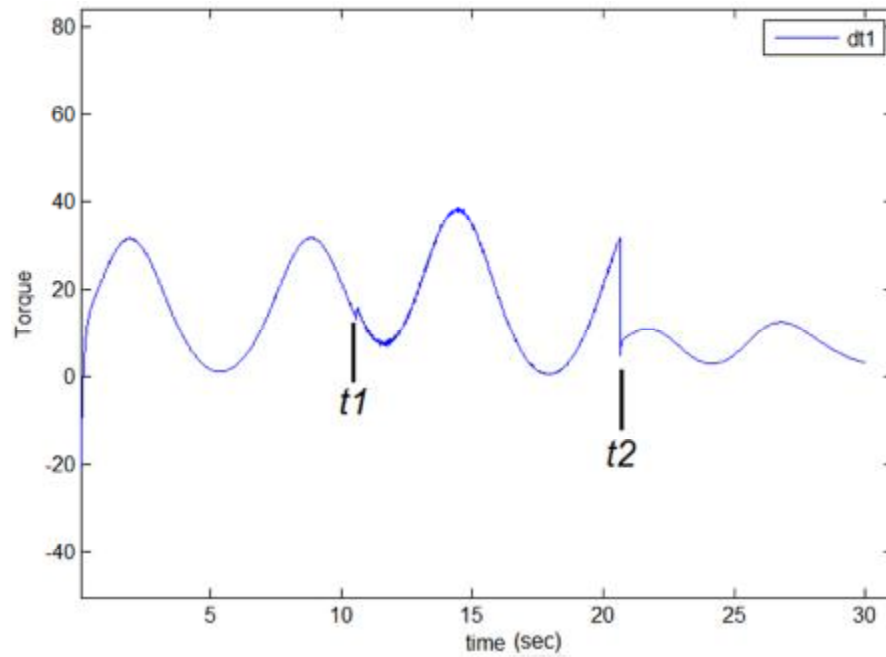


Figure 4.11. Model 2 HAMM Joint 1 Actuator Torques

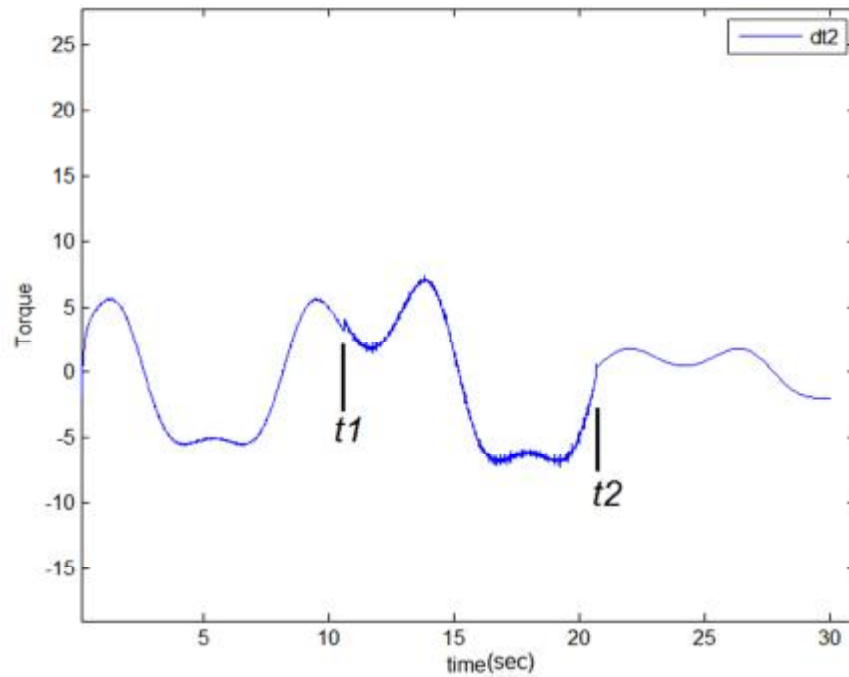


Figure 4.12. Model 2 HAMM Joint 2 Actuator Torques

Orthesis joint 1 and joint 2 actuator torques obtained from the same simulations are shown in Figure 4.13 and 4.14.

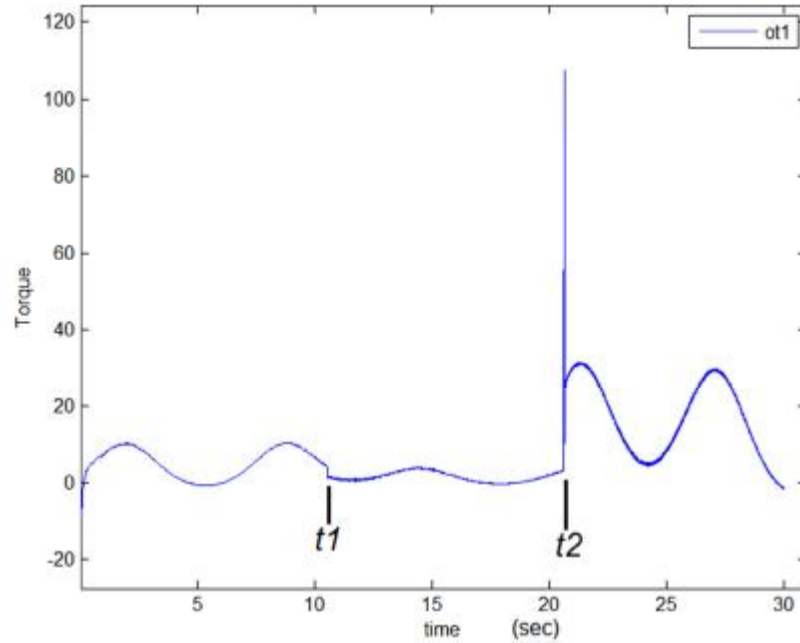


Figure 4.13. Model 2 Orthesis Joint 1 Actuator Torques

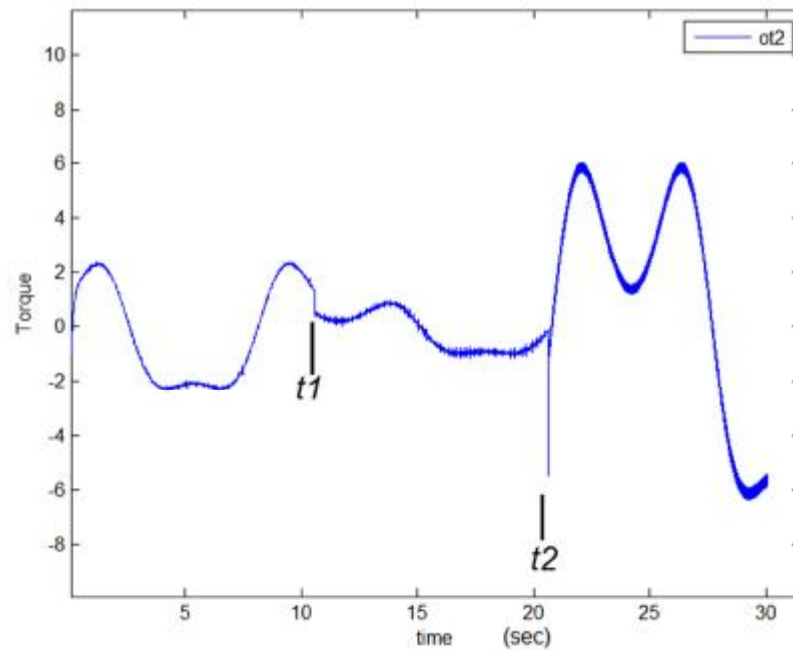
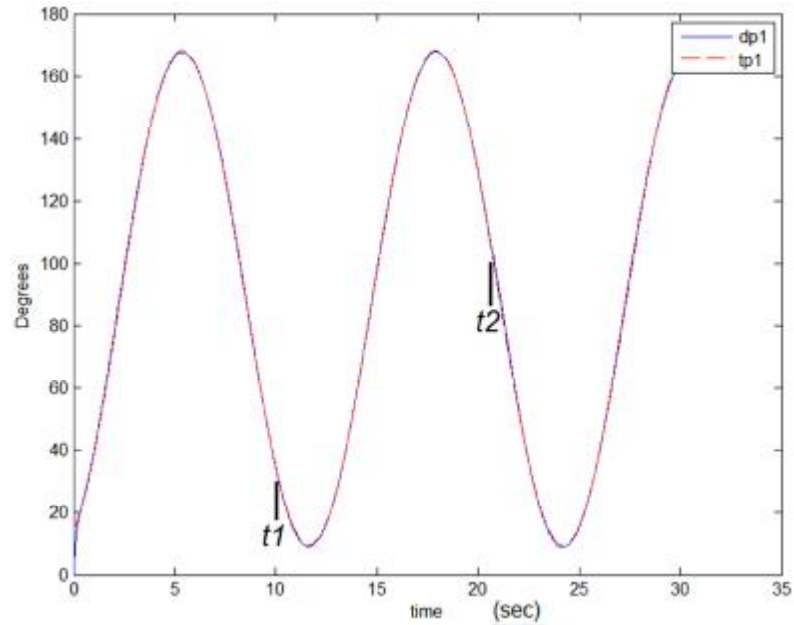
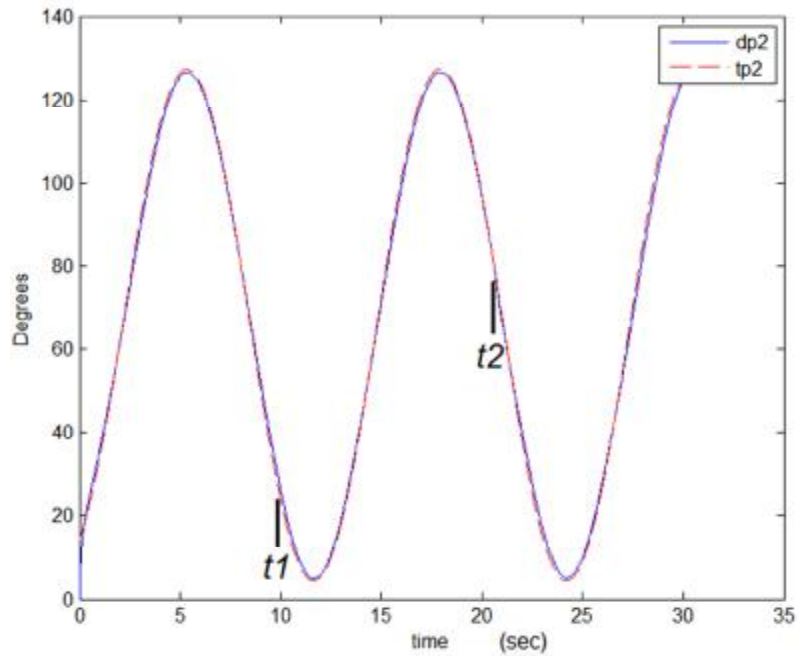


Figure 4.14. Model 2 Orthesis Joint 2 Actuator Torques

The desired trajectories for both joints and HAMM joint positions in simulation are shown in Figure 4.15.



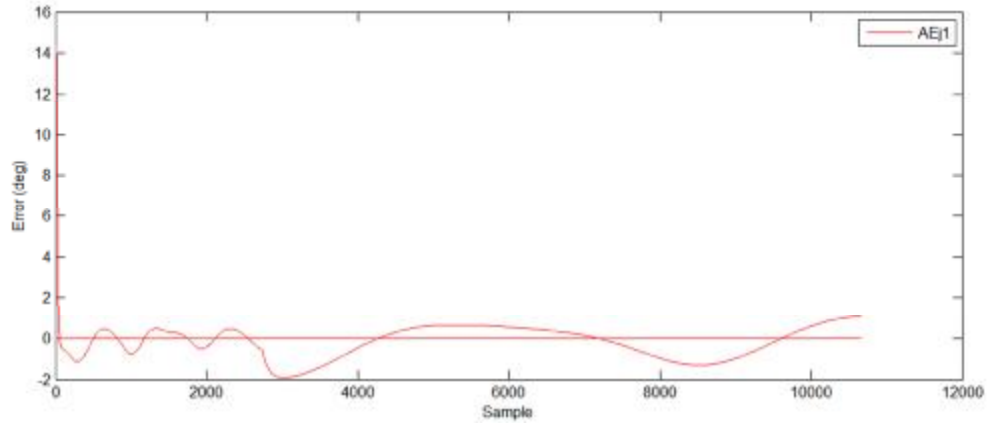
(a) Model 2 HAMM Joint1 (dp1) Position And Desired Trajectory (tp1) For Joint 1



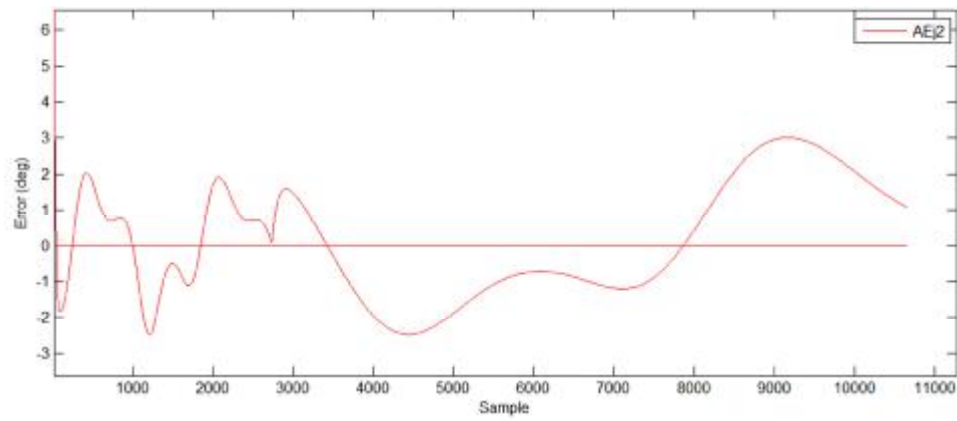
(b)

Figure 4.15. Model 2 (a) HAMMJoint1 (dp1) Position And Desired Trajectory (tp1) For Joint 1 (b) HAMM Joint 2 (dp2) Position And Desired Trajectory (tp2) For Joint 2

AE for HAMM joint positions in simulation 2 are given in Figure 4.16.



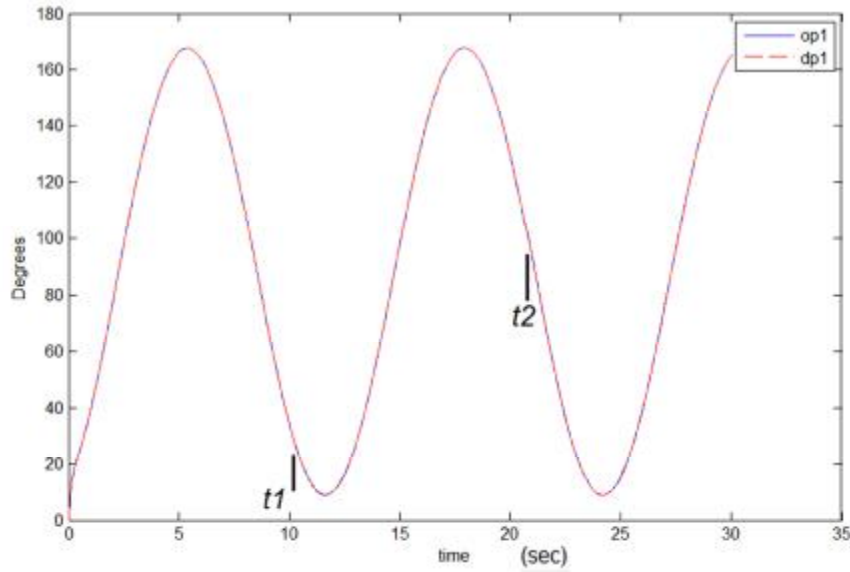
(a) AE for HAMM Joint1 Position



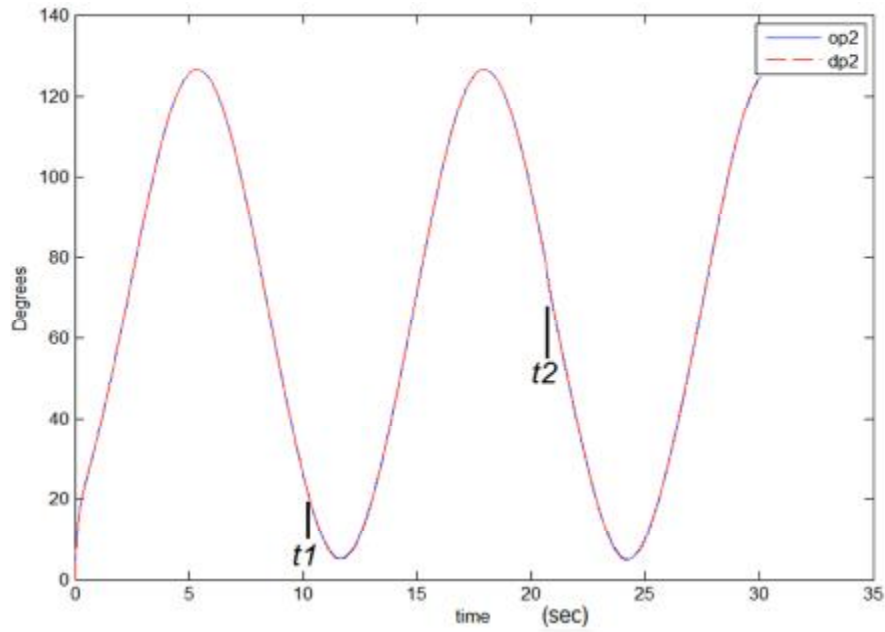
(b)

Figure 4.16. (a) AE for HAMM Joint1 Position (b) AE for HAMM Joint2 Position

HAMM and orthosis joint positions during simulation are shown in Figure 4.17.



(a) Model 2 HAMM And Orthesis Joint1 (dp1, op1) Position



(b)

Figure 4.17. Model 2 (a) HAMM And Orthesis Joint1 (dp1, op1) Position (b) HAMM And Orthesis Joint2 (dp2, op2) Position

Analysing the results obtained from 2nd model simulation, HAMM joint torques increase at resistive mode, and they decrease at assistive mode. From the Figure 4.15, the position error occurred in HAMM 2nd joint position while switching to resistive mode in 1st model, did not take place in 2nd model.

4.3. Simulation Results of Model 3

The control of orthosis was made by using DM1 and the control of HAMM was made by using DM2 in model 3. The orthosis and HAMM joint 1 actuator torque values of the simulation of 3rd model are given in Figure 4.18. The orthosis was operated in normal mode for first 10 seconds during simulation. The orthosis had been at assistive mode 2nd level starting from t_1 time instant. The orthosis switched to resistive mode 2nd level after t_2 time instant. The orthosis switched to assistive mode 3rd level after t_3 time instant. t_1 , t_2 and t_3 time instants are shown in Figures.

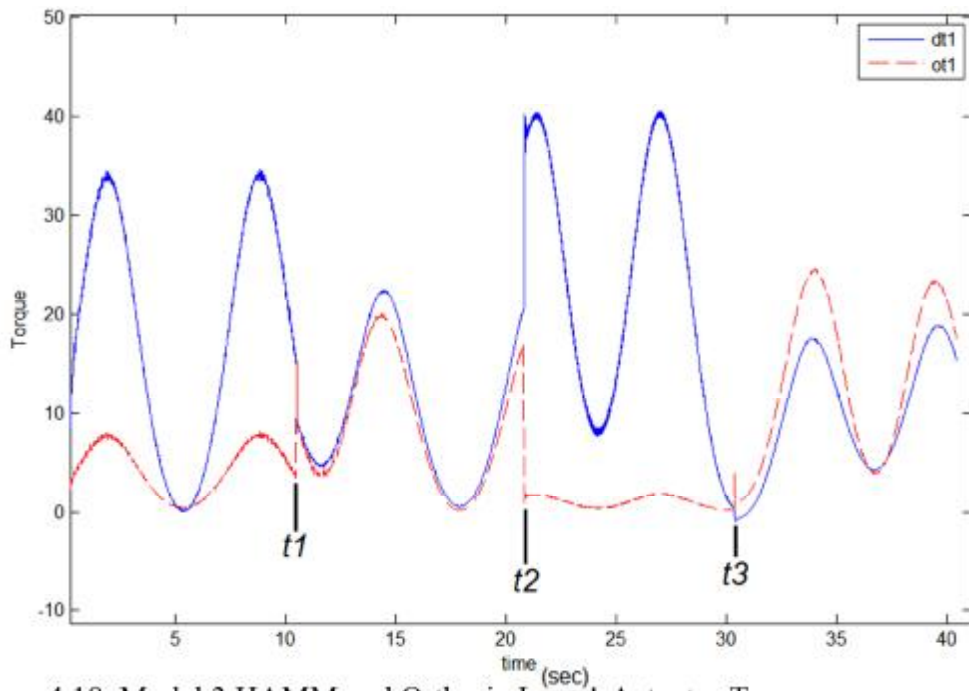


Figure 4.18. Model 3 HAMM and Orthosis Joint 1 Actuator Torques

Orthosis and HAMM joint 2 actuator torques obtained from the simulation are shown in Figure 4.19.

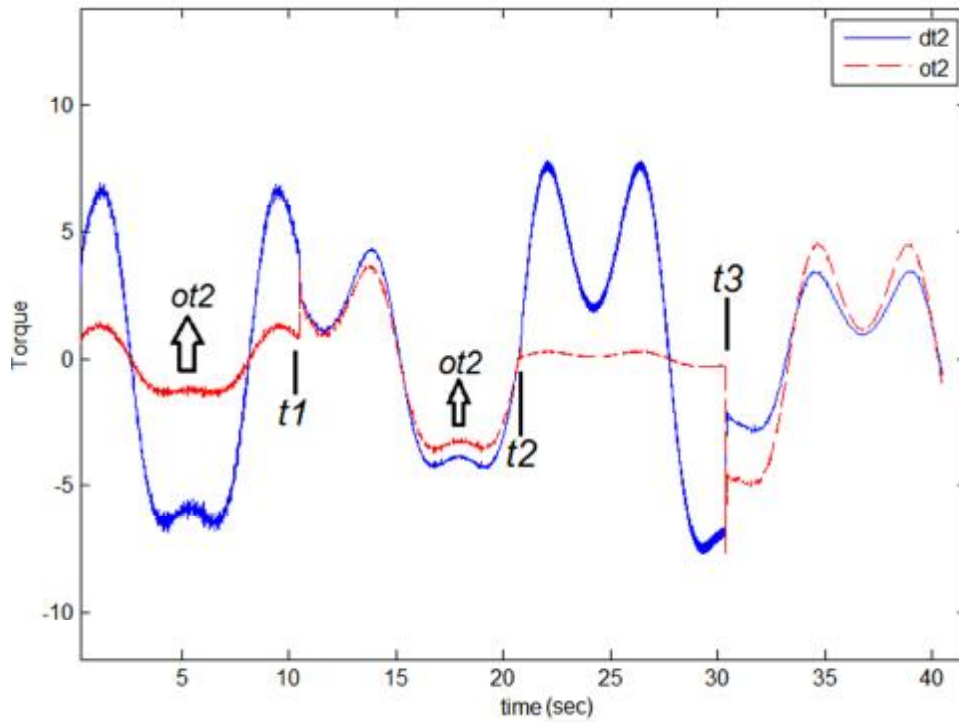
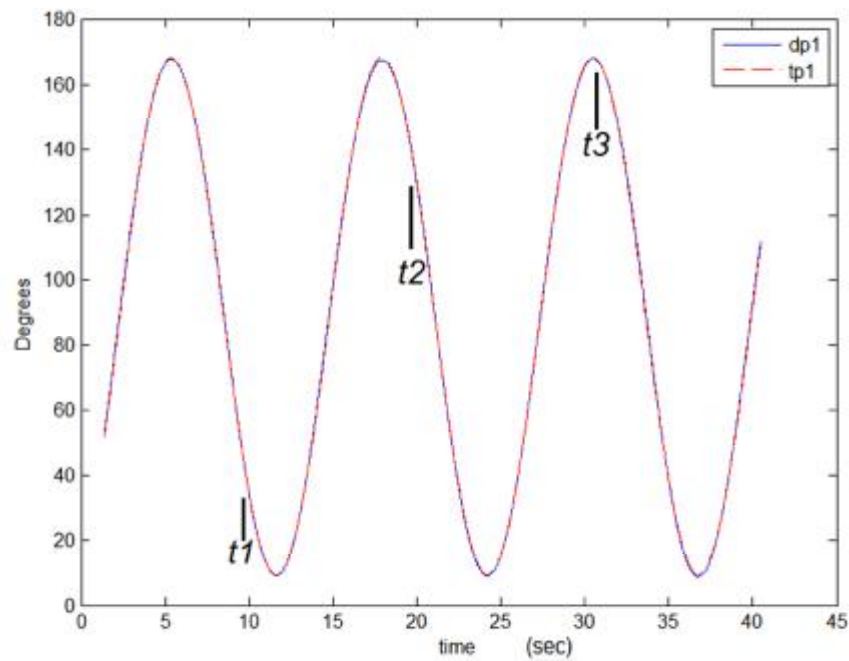
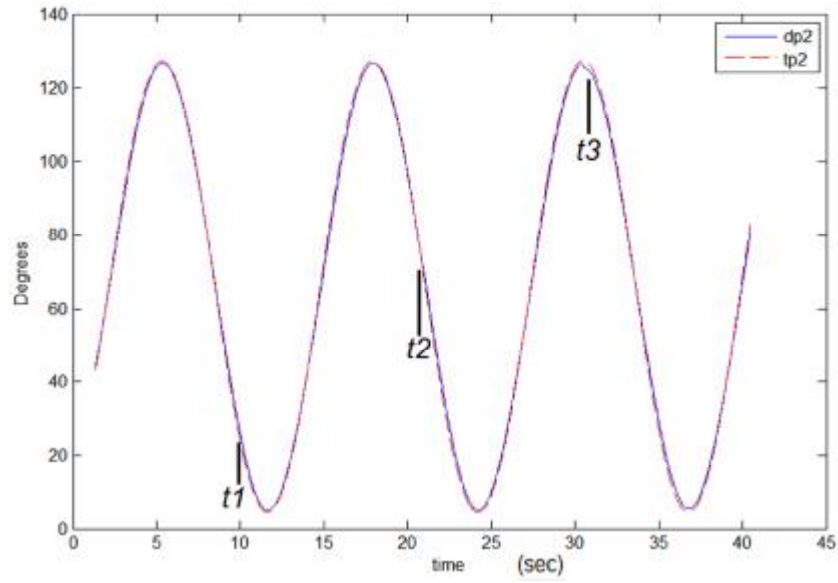


Figure 4.19. Model 3 HAMM and Orthosis Joint 2 Actuator Torques

The desired trajectories for both joints and HAMM joint positions during simulation are shown in Figure 4.20.



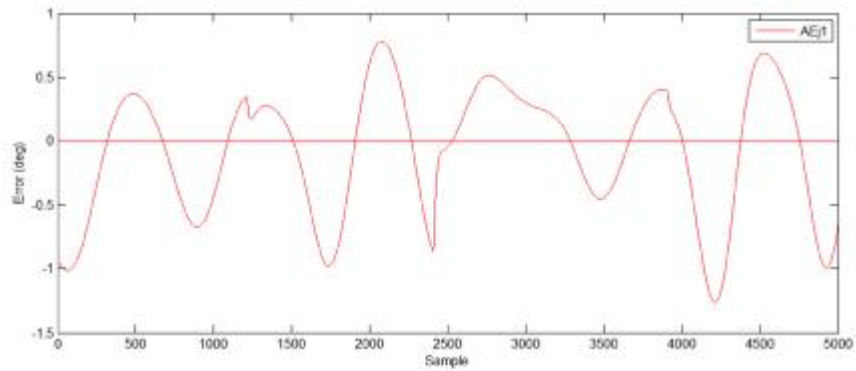
(a)



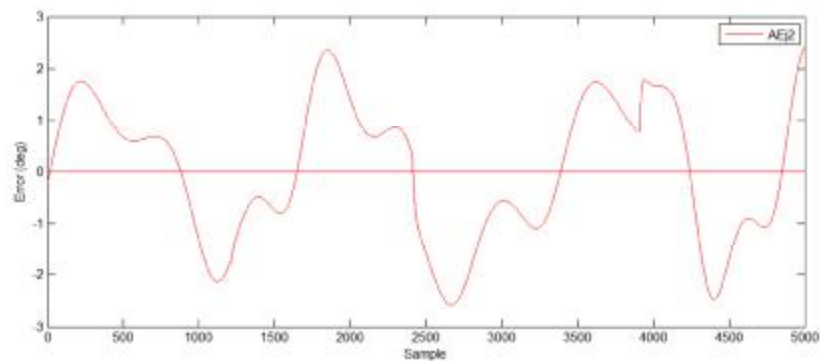
(b)

Figure 4.20. Model 3 HAMM Joint Positions and Desired Trajectories

AE for HAMM joint positions in simulation 3 are given in Figure 4.21.



(a) AE for HAMM Joint1 Position



(b)

Figure 4.21. (a) AE for HAMM Joint1 Position (b) AE for HAMM Joint2 Position

HAMM and othesis joint positions during this simulation are shown in Figure 4.22.

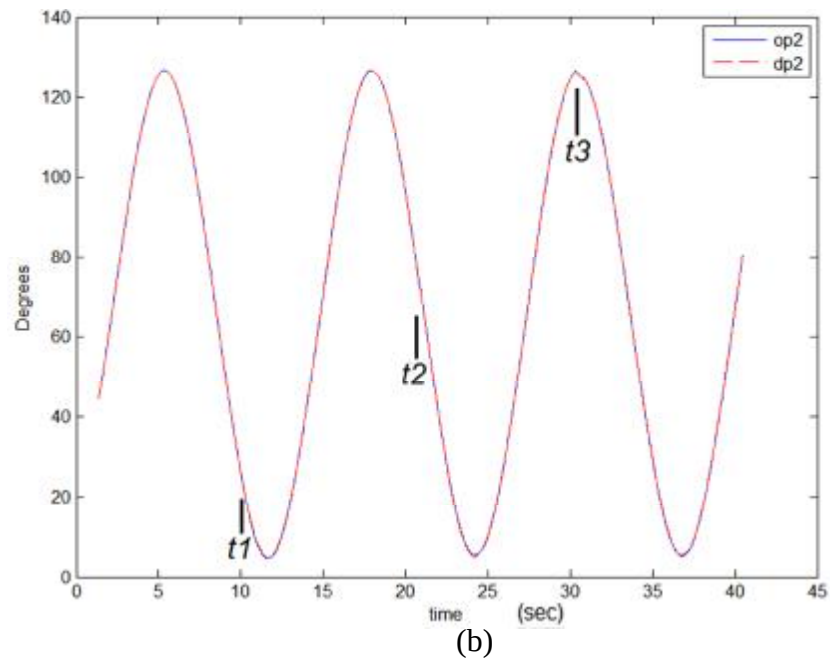
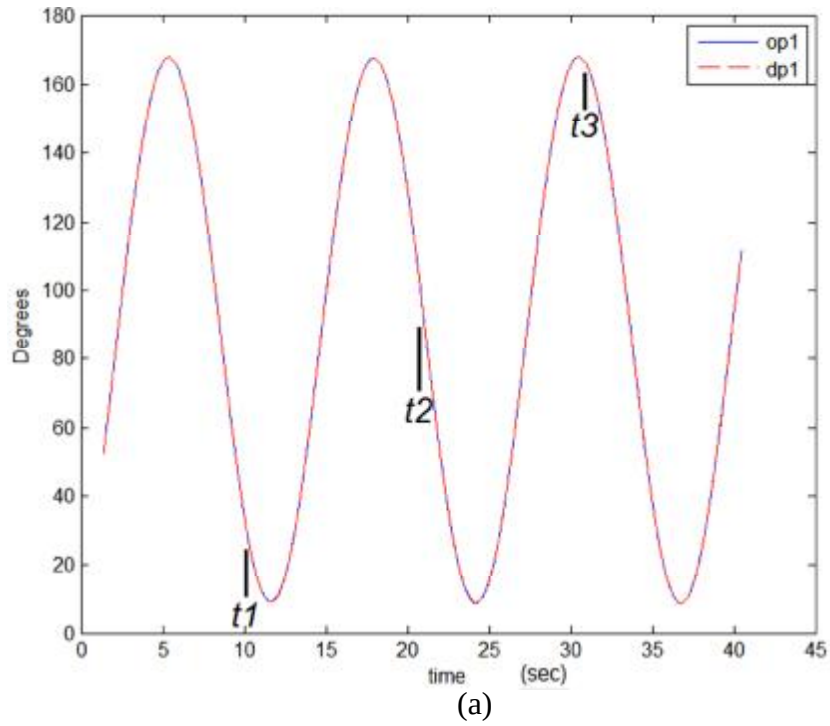


Figure 4.22. Model 3 HAMM and Orthosis Joint Positions

As analysed the results obtained from 3rd model simulation, HAMM joint torques increase at resistive mode, and they decrease at assistive mode which shows the rehabilitation modes work correctly. In Figures 4.20 and 4.22, error occurred in both of HAMM joint positions while switching from resistive mode to assistive mode (t_3).

4.4. Simulation Results for Model 4

The control of orthosis was made by using DM2 and the control of HAMM was made by using DM1 in model 4. The orthosis and HAMM joint 1 actuator torque values of the simulation of 4th model are given in Figure 4.23. The orthosis was operated in normal mode for first 10 seconds during simulation. The orthosis had been at resistive mode 4th level starting of t_1 time instant. The orthosis switched to assistive mode 5th level after t_2 time instant. The orthosis switched to resistive mode 2nd level after t_3 time instant. t_1 , t_2 and t_3 time instants are shown in Figures.

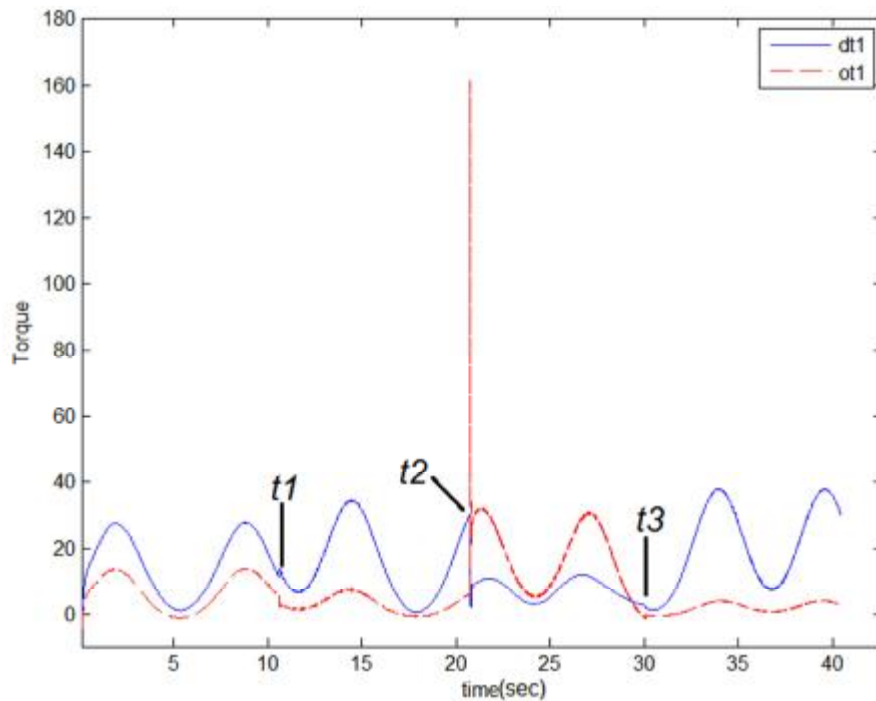


Figure 4.23. Model 4 HAMM and Orthosis Joint 1 Actuator Torques

Orthesis and HAMM joint 2 actuator torques obtained from the simulation are shown in Figure 4.24.

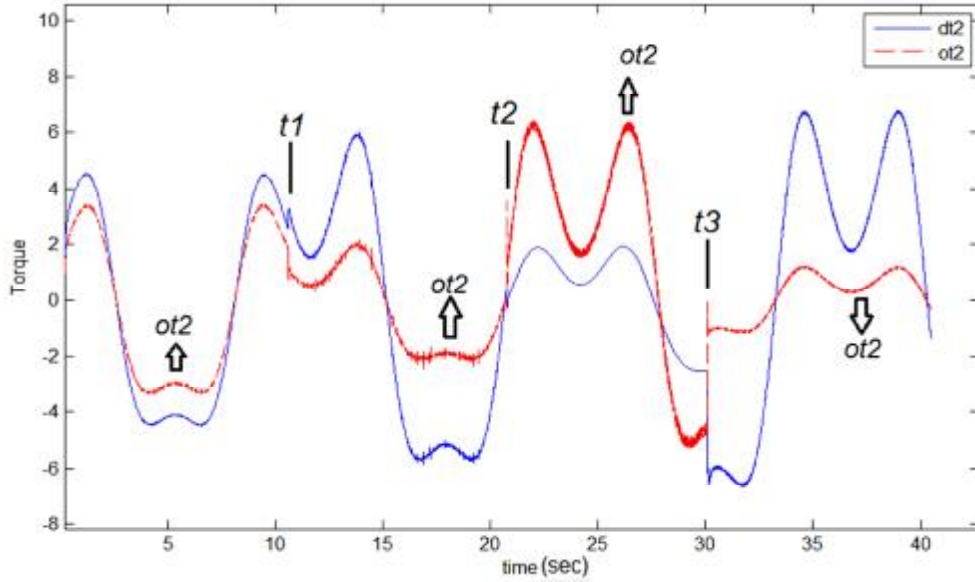
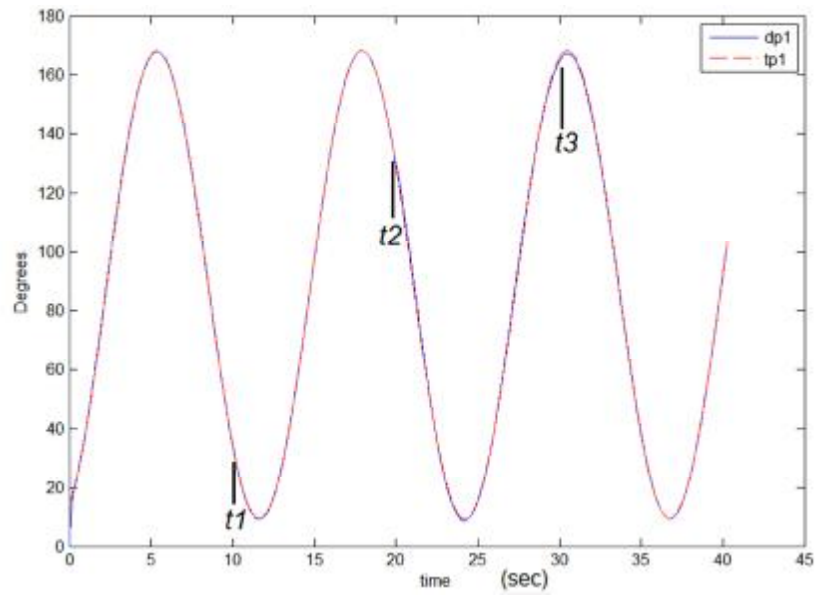
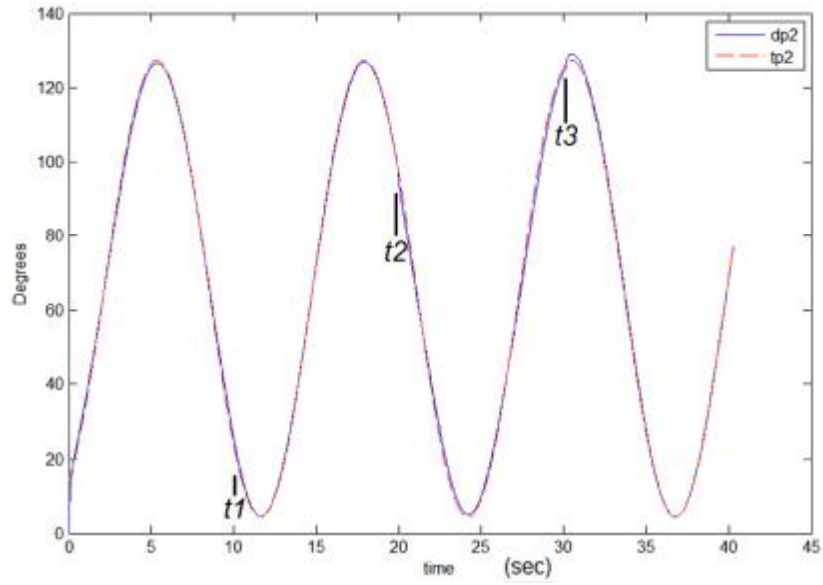


Figure 4.24. Model 4 HAMM and Orthesis Joint 2 Actuator Torques

The desired trajectories for both joints and HAMM joint positions in simulation 4 are shown in Figure 4.25.



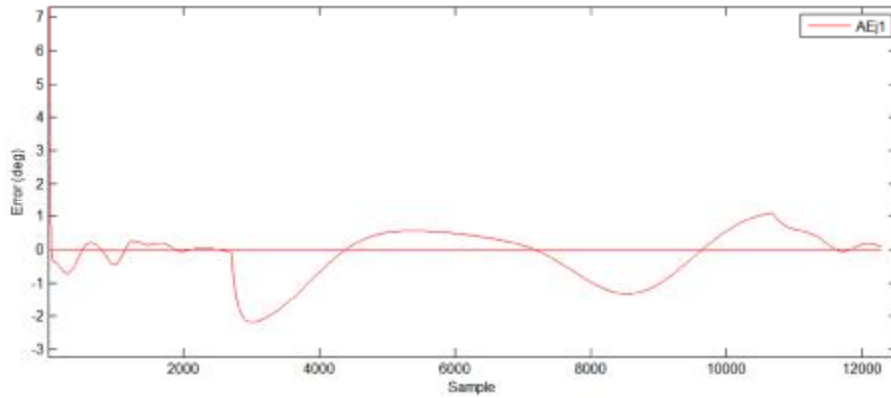
(a)



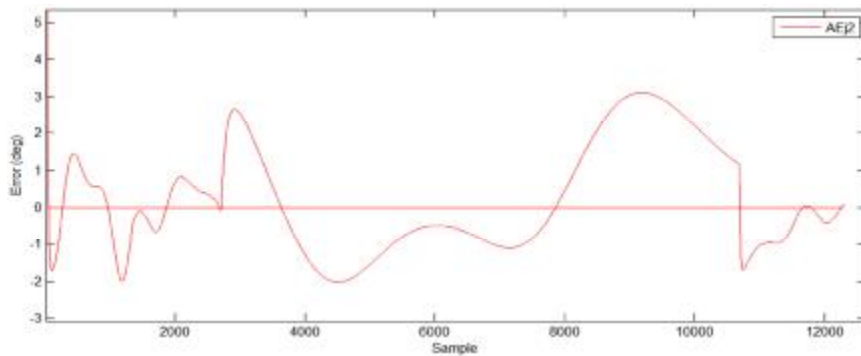
(b)

Figure 4.25. Model 4 HAMM Joint Positions and Desired Trajectories

AE for HAMM joint positions in simulation 4 are given in Figure 4.26.



(a) AE for HAMM Joint1 Position



(b)

Figure 4.26. (a) AE for HAMM Joint1 Position (b) AE for HAMM Joint2 Position

HAMM and orthesis joint positions during simulation are shown in Figure 4.27.

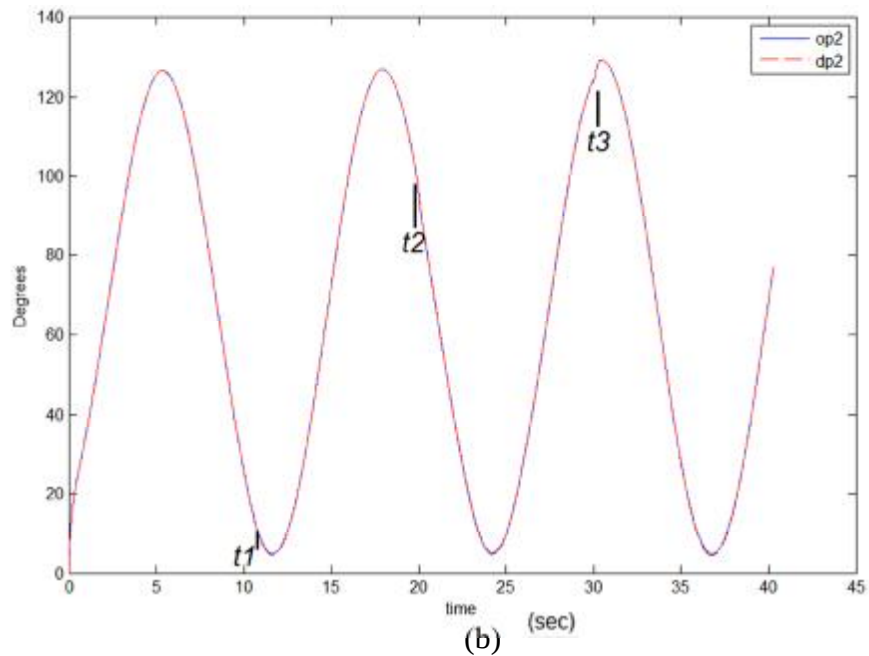
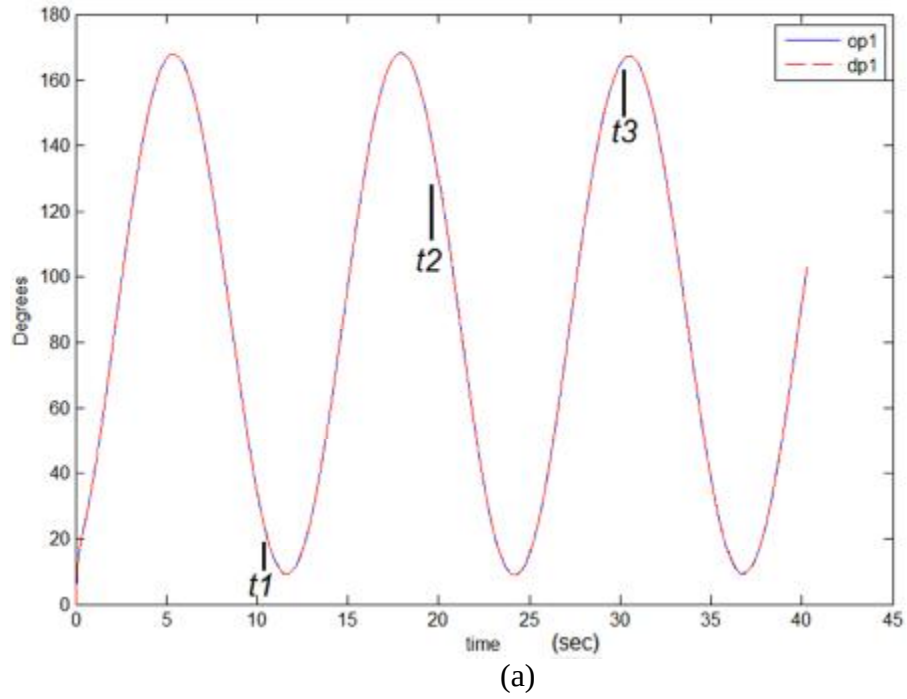
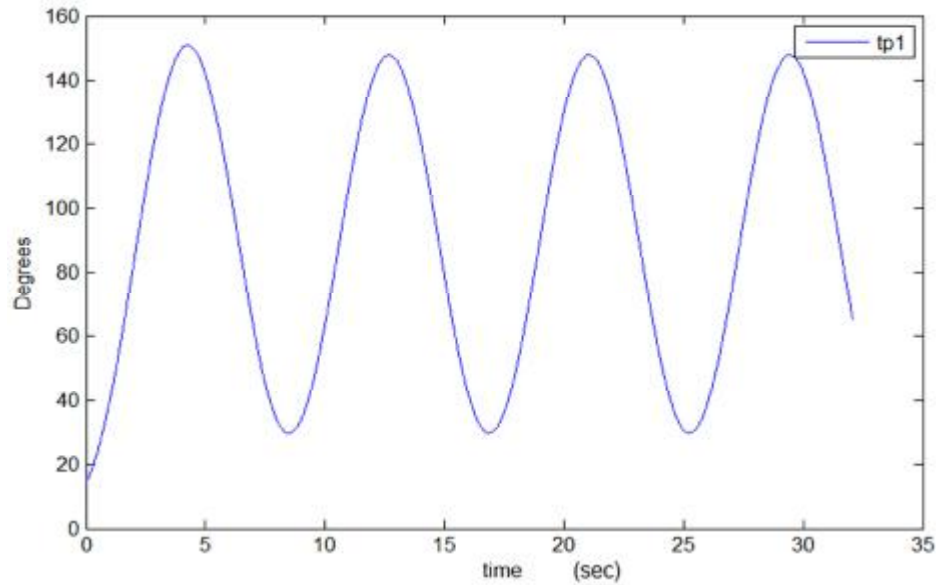


Figure 4.27. Model 4 HAMM and Orthesis Joint Positions

When analysed the results obtained from 4th model simulation, HAMM joint torques increase at resistive mode and they decrease at assistive mode. In Figure 4.25, error occurred in both of HAMM joint positions while switching from assistive mode to resistive mode (t_3).

4.5. Simulation Results of Model 3 with Second Trajectory

The 3rd model which used DM1 for the controlling orthosis and DM2 for the controlling HAMM were tested with a different desired trajectory to observe the performance of designed controller. The second desired trajectory for both joints is shown in Figure 4.28.



(a)

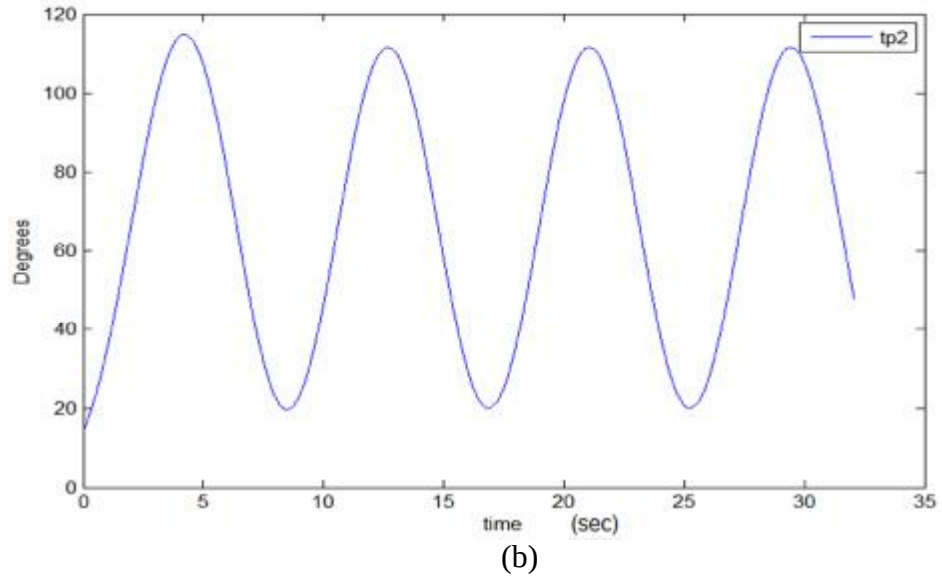


Figure 4.28. 2nd Desired Trajectory for Joint 1 (a) and Joint 2 (b)

The orthosis was operated in normal mode for first 10 seconds with second desired trajectory during simulation of the 3rd model. The orthosis had been at resistive mode 3rd level starting of t_1 time instant. The orthosis switched to assistive mode 4th level after t_2 time instant. t_1 and t_2 time instants are shown in Figures.

Orthesis and HAMM joint 1 actuator torques obtained from the simulation are shown in Figure 4.29.

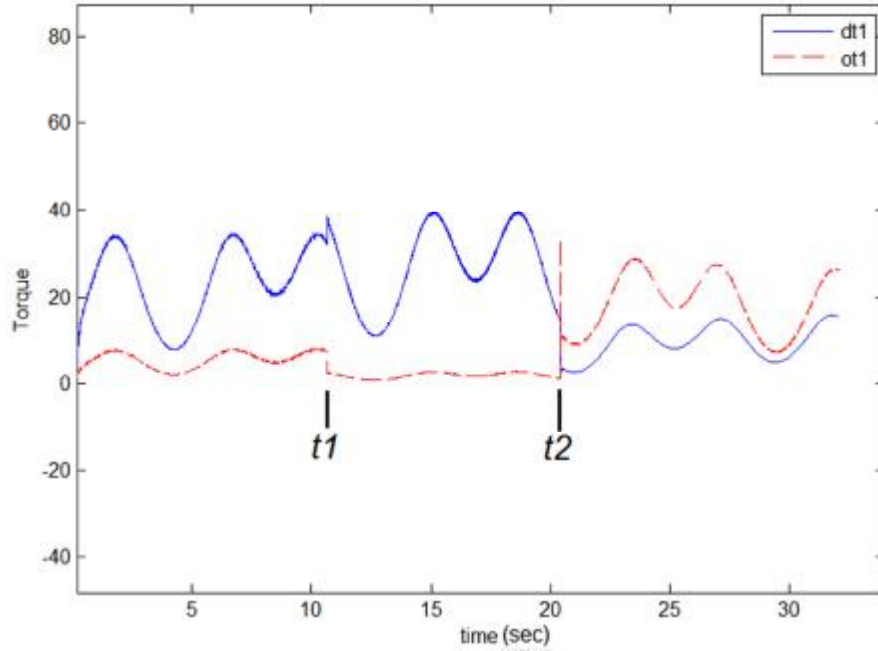


Figure 4.29. Orthesis and HAMM Joint 1 Actuator Torques with 2nd Desired Trajectory

Orthesis and HAMM joint 2 actuator torques obtained from the simulation are shown in Figure 4.30.

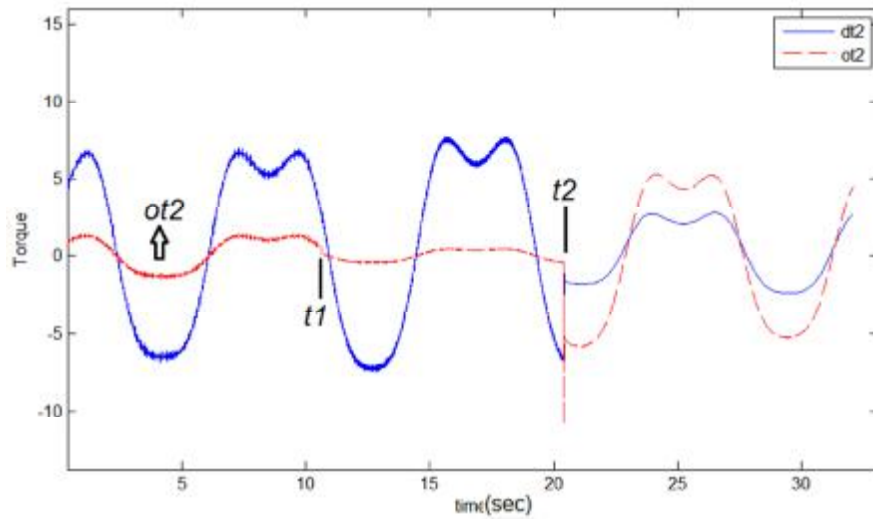


Figure 4.30. Orthesis and HAMM Joint 2 Actuator Torques with 2nd Desired Trajectory

The desired trajectories for both joints and HAMM joint positions in this simulation are shown in Figure 4.31.

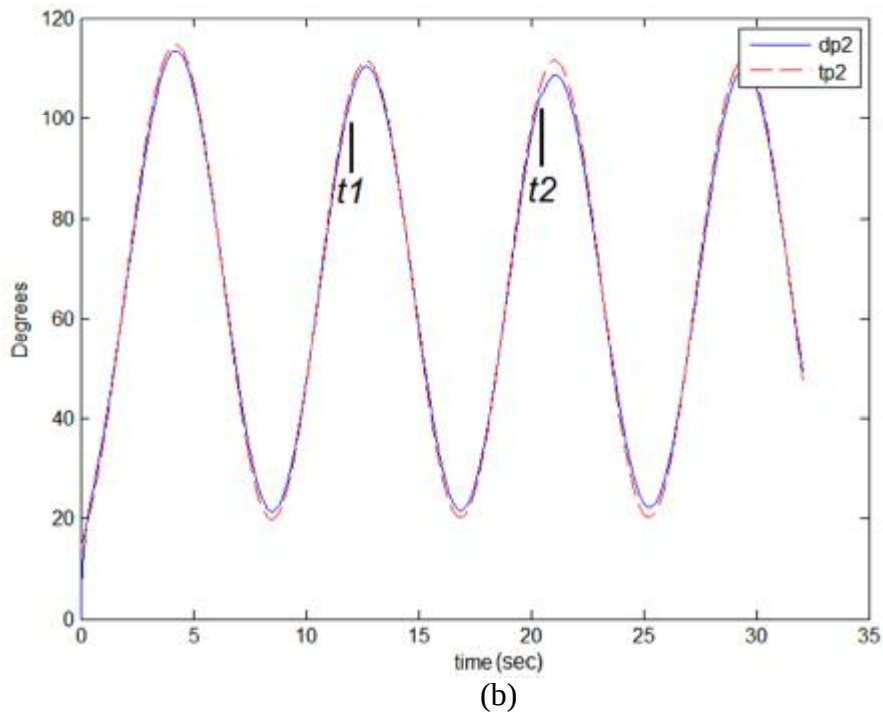
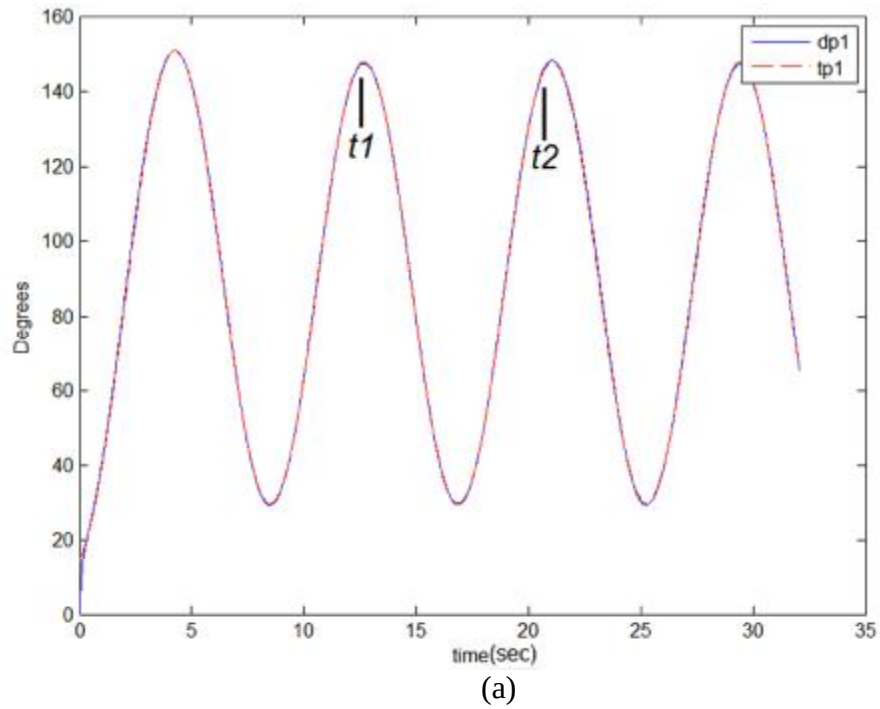
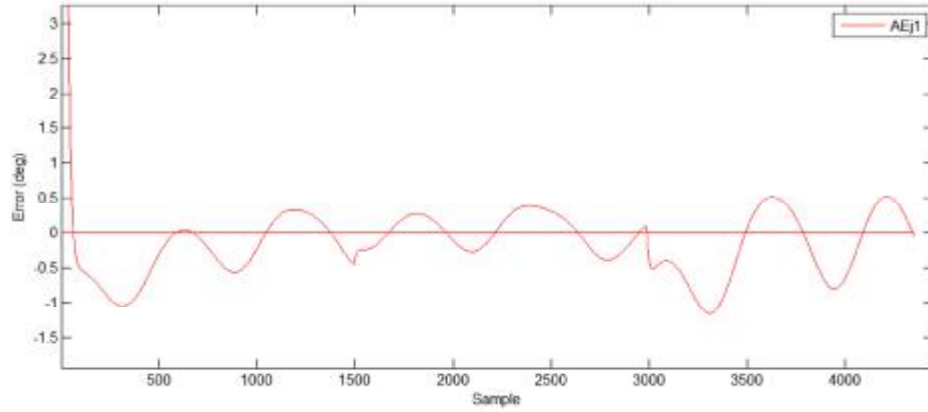
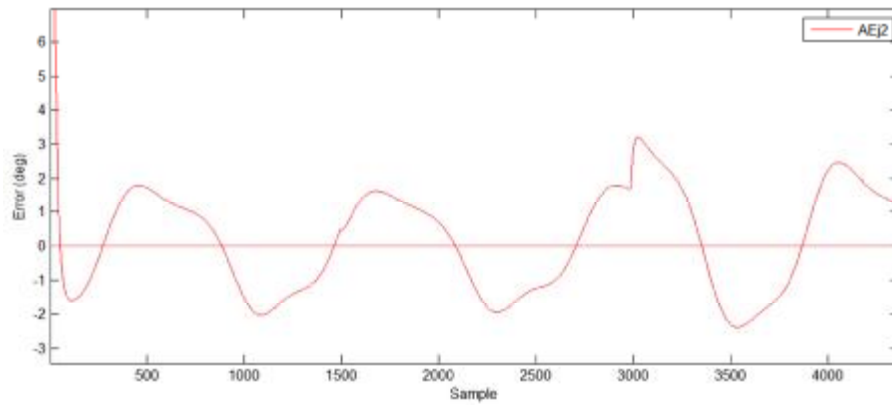


Figure 4.31. HAMM Joint Positions and 2nd Desired Trajectory

AE for HAMM joint positions in simulation 5 are given in Figure 4.32.



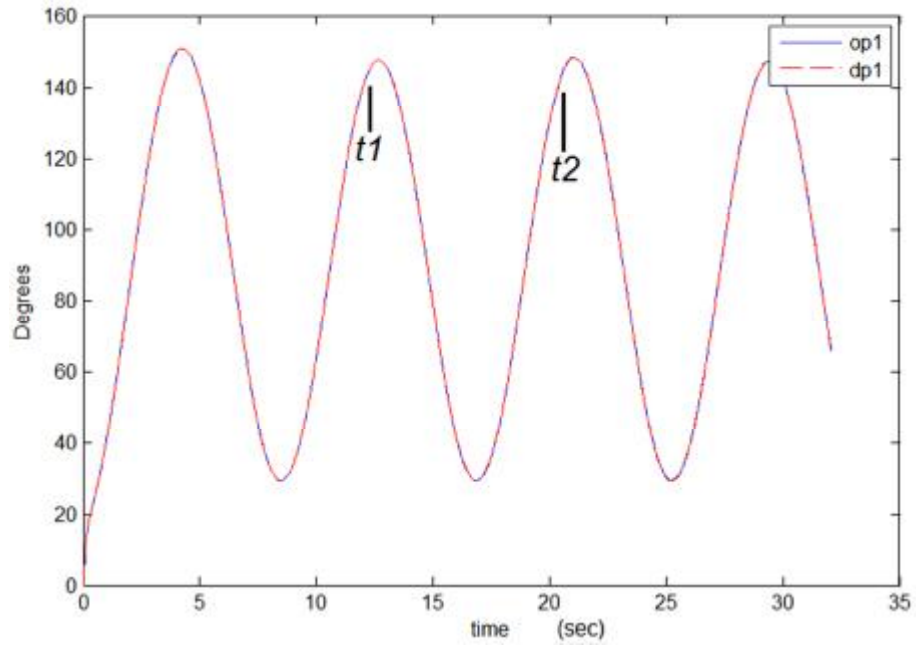
(a) AE for HAMM Joint1 Position



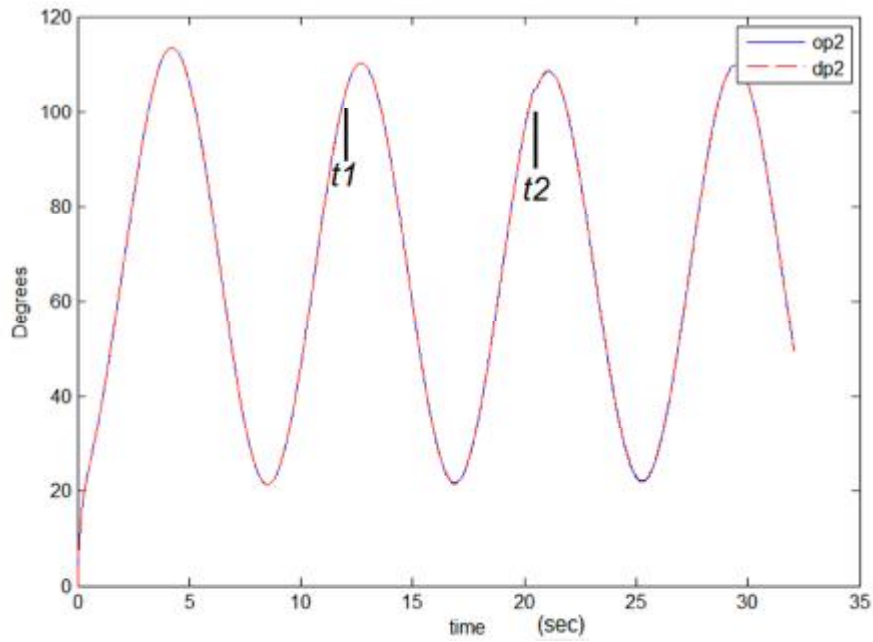
(b)

Figure 4.32. (a) AE for HAMM Joint1 Position (b) AE for HAMM Joint2 Position

HAMM and orthosis joint positions during this simulation are shown in Figure 4.33.



(a)



(b)

Figure 4.33. HAMM and Orthesis Joint Positions in Simulation with 2nd Desired Trajectories

Analysing the results of Figures 4.29 and 4.30, HAMM joint torques increase at resistive mode of orthesis, and they decrease at assistive mode. This indicates that, rehabilitation modes works independently from trajectory. In Figure 4.31, HAMM

followed second desired trajectory during simulation. But the increase of trajectory frequency also increased the errors occurred in HAMM 2nd joint position as shown in Figure 4.31.

4.6. Effect of Spring Constant to Results

The simulation with 4th model which used DM2 for the control of orthosis and DM1 for the control of HAMM was repeated by changing the spring constant value of Body Spring & Damper Block which connecting the orthosis and HAMM. It is aimed to observe the effect of spring constant to the simulation. The simulation was operated at normal mode for the first 10 seconds. The orthosis was at assistive mode 4th level starting of t_1 time instant. Spring constant parameter of Spring & Damper Block was 95×10^3 . Spring constant value was lowered to 5×10^4 at the moment of t_2 . Spring constant value was lowered to 5×10^3 at t_3 time instant. Spring constant value was adjusted to 2×10^5 at t_4 time instant. t_1 , t_2 , t_3 , and t_4 time instants are given in Figures.

Orthesis and HAMM joint 1 actuator torques obtained from the simulation are shown in Figure 4.34.

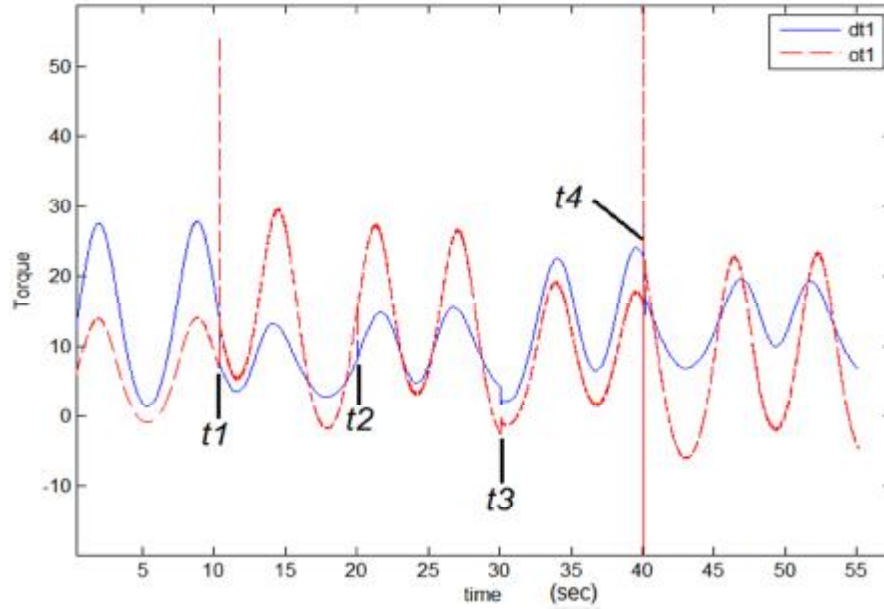


Figure 4.34. Simulation 6 HAMM and Orthesis Joint 1 Actuator Torques

Orthesis and HAMM joint 2 actuator torques obtained from the simulation are shown in Figure 4.35.

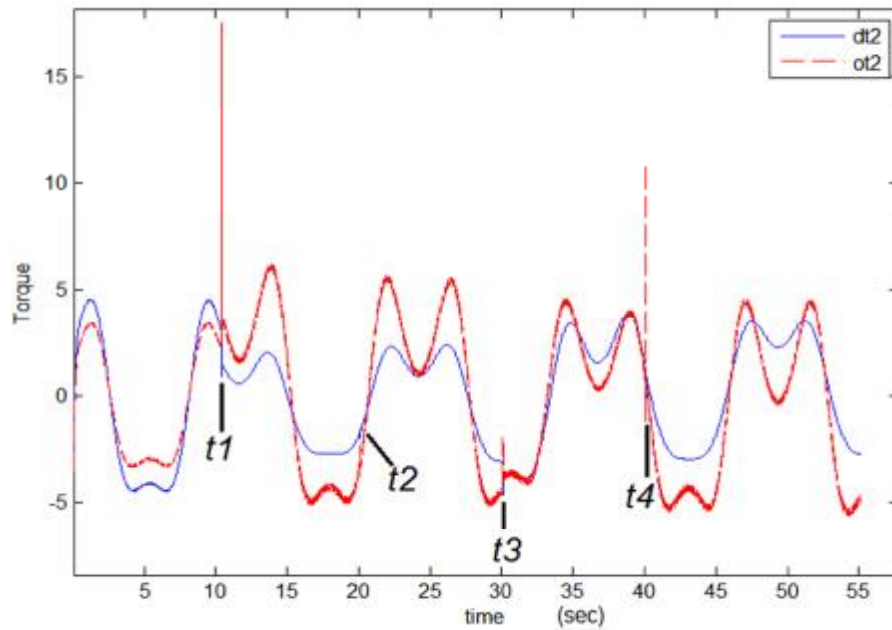
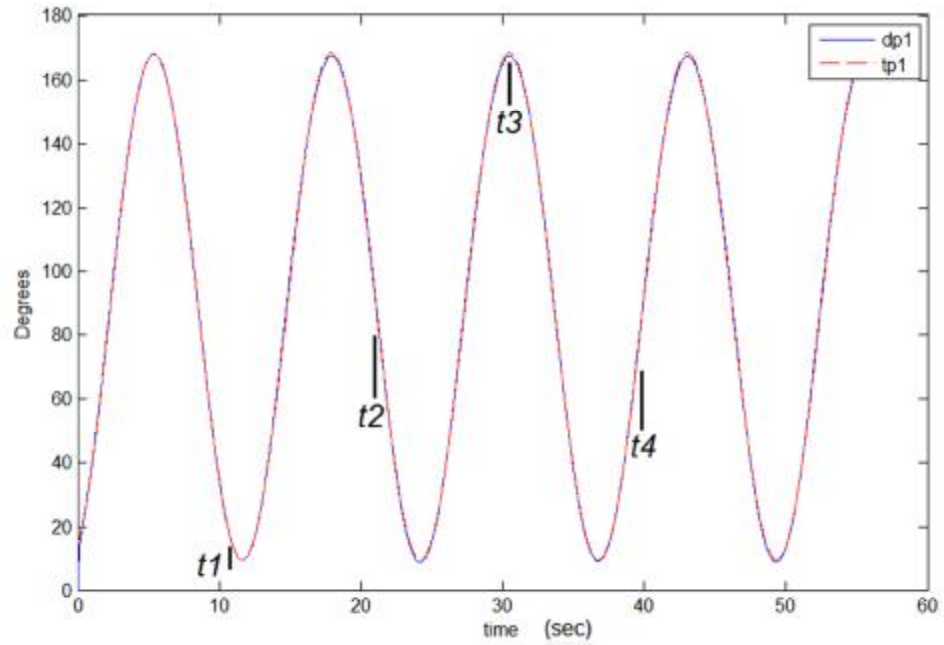
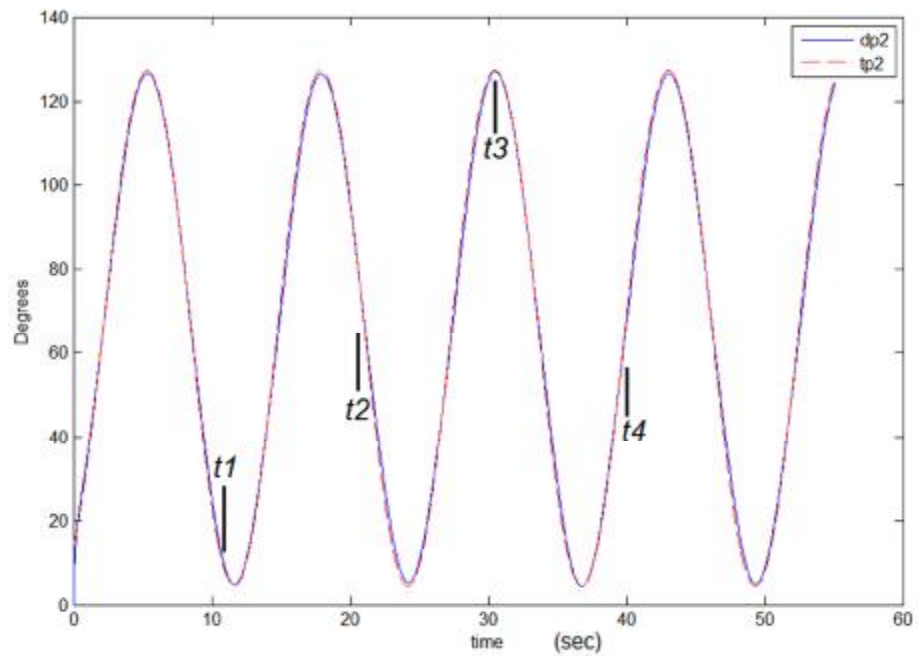


Figure 4.35. Simulation 6 HAMM and Orthesis Joint 1 Actuator Torques

The desired trajectories for both joints and HAMM joint positions in this simulation are shown in Figure 4.36.



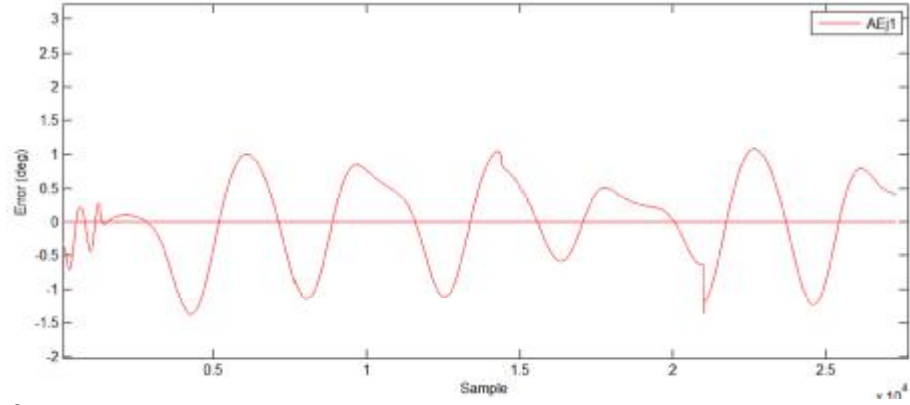
(a)



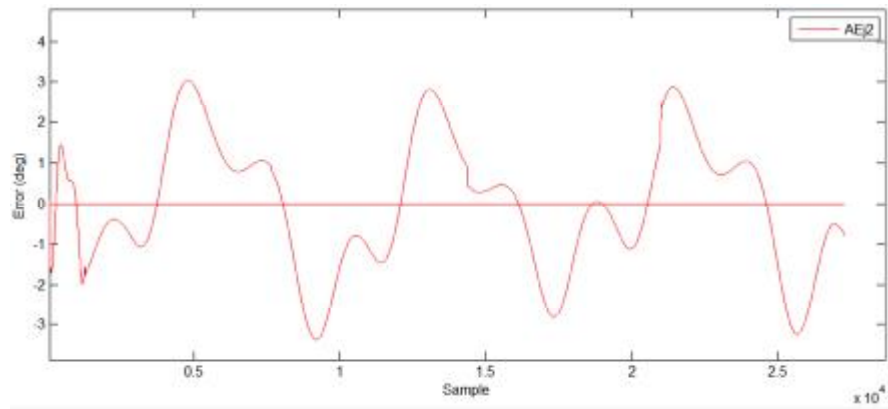
(b)

Figure 4.36. Simulation 6 Desired Trajectories and HAMM Joint Positions

AE for HAMM joint positions in simulation 6 are given in Figure 4.37.



(a) AE for HAMM Joint1 Position



(b)

Figure 4.37. (a) AE for HAMM Joint1 Position (b) AE for HAMM Joint2 Position

HAMM and othesis joint positions during simulation are shown in Figure 4.38.

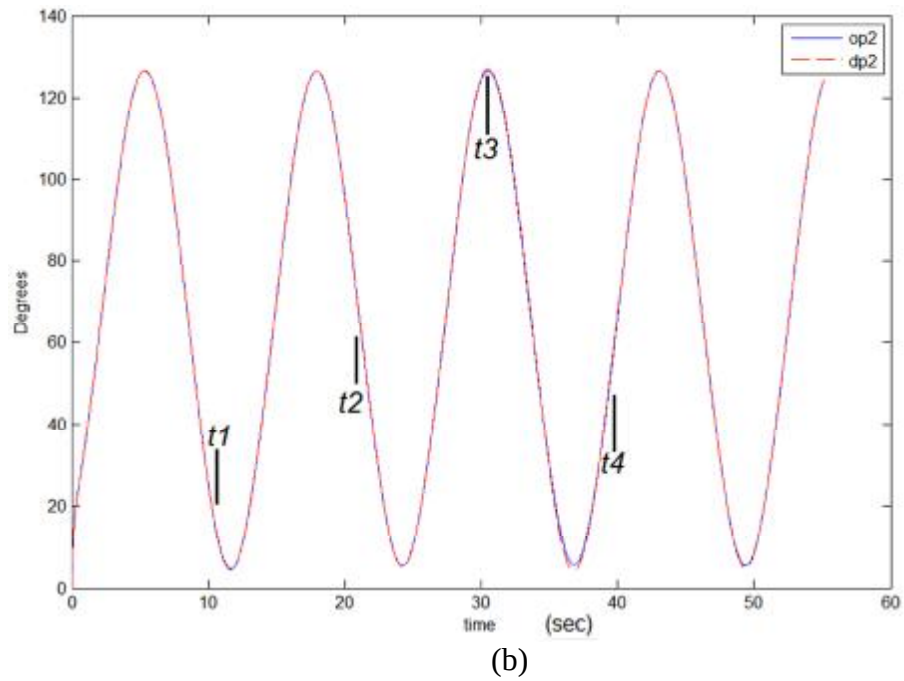
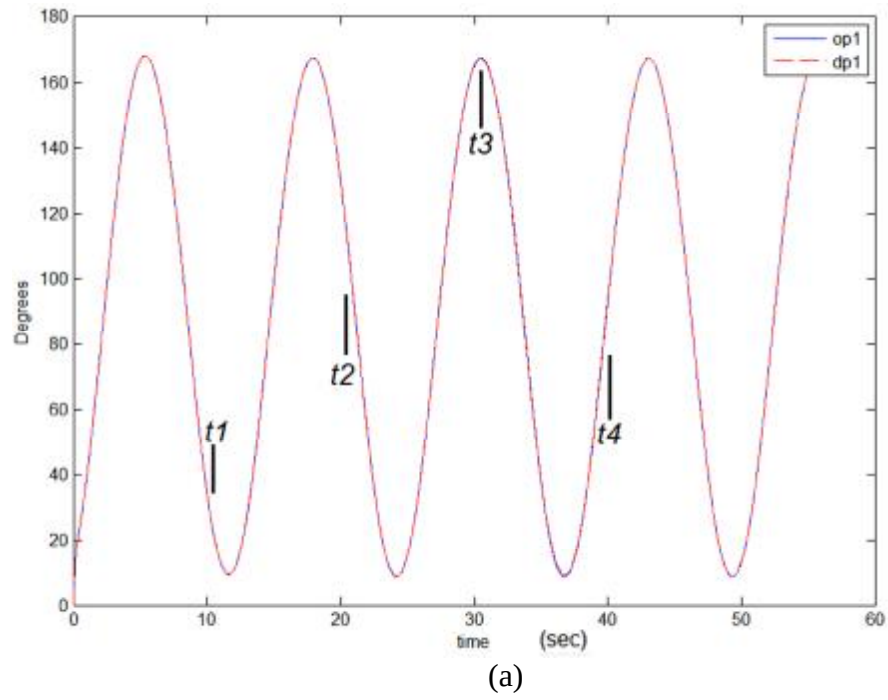


Figure 4.38. Simulation 6 HAMM and Orthesis Joint Positions

When analysed the results of Figures 4.34 and 4.35, HAMM joint torques increase when the tightness of the connection of HAMM and orthosis (t_2 and t_3) was lessened by decreasing spring constant value although there were not any mode transitions. In this situation, it is understood that the decrease of tightness of the connection reduces the HAMM and orthosis interaction. Tightness increase during t_4 had no significant effect on joint torques. In Figures 4.36 and 4.38, the changes made at the spring constant after simulation started caused errors at joint positions at the time of changes.

4.7. Stability of The System

Joint variables average position errors are given in Figures 4.8, 4.9, 4.16, 4.21, 4.26, 4.32 and 4.37 for HAMM and orthosis. When the Figures are analyzed it is observed that Average Position Error (AE) values increase during mode transitions when orthosis or HAMM joint torques either decrease or increase. This happens due to the fact that interaction forces between orthosis and HAMM during mode transitions create disturbance effect in the system.

However this effect is regulated by the control system and steady state error reaching towards zero indicates the stability of the system. The position error is bounded around zero. Globally asymptotic stable behavior is observed.

4.8. Discussion

Results show that, the controllers implemented the assistive and resistive modes, mode levels and mode transitions successfully. Dynamic model 1 was better than dynamic model 2 at trajectory tracking. But, it was observed that position errors which didn't occur during mode transitions in the model which used only dynamic model 2 occurred in all models which used dynamic model 1. Controller was also tested with another trajectory, but the increase of trajectory frequency increased the errors of second joint position. Body Spring & Damper Block spring constant value was changed and its effect on results was tested during simulation. According to

results, the change of spring constant value did not affect mode transitions, but the change of connection tightness between orthosis and HAMM affected the performance of rehabilitation modes.

5. CONCLUSION AND FUTURE WORKS

In this thesis, the design of a multi-mode model based controller and a robotic arm orthosis with two DoF in connection to a human arm mechanical model is presented.

Mechanical properties of the designed orthosis system were obtained from Solidworks CAD drawings of the design. PID Computed-torque controller was selected as a control method because of its success in structures which mechanical properties were known properly. Dynamic model of the orthosis was created with two different methods and its performance was analyzed comparatively. The designed system was implemented in simulation for compliant operation of robotic arm orthosis and human arm model.

The controller is tested for multi-mode operation and its performance is evaluated for “Assistive” and “Resistive” rehabilitation modes. Analysing the results of simulation, the designed controller is implemented in Assistive and Resistive modes for both dynamic models and mode transitions successfully. Errors were encountered during mode transitions with the models which used dynamic model 1. The tests yielded encouraging results for future works.

For the future works, the performance of the dynamic model based controller will be compared to the performance of the Hill-muscle model based controller. Besides, a high level controller will be added to the control structure. Eventually, it is aimed to test the controller performance in a robotic orthosis hardware.

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CURRICULUM VITAE

Erkan ÖDEMİŞ was born in Şanlıurfa, Turkey in 1988. He received his B.S. degree in Electrical and Electronics Engineering Department from Çukurova University in 2012. After completion his B.S. education, he started MSc education in Electrical and Electronics Engineering Department in Çukurova University in 2012. He has been working as a Research Assistant in Biomedical Engineering Department of Çukurova University since 2012. His research areas are Robotics, Rehabilitation Robotics and Exoskeleton Robotics.

APPENDIX

Orthesis Dynamic Model 1 Equations

```
function [t1,t2]= fcn( da1,da2,ap1, ap2, av1,av2,dp1,dp2,dv1,dv2,E1,E2)

ep1 = dp1-ap1;
ep2 = dp2-ap2;
ev1 = dv1-av1;
ev2 = dv2-av2;

m1 = 2.4 ; m2 = 2.02 ; l1 = 0.32 ; l2 = 0.26 ; g = 9.8 ; Lc1 = 0.128 ; Lc2 =
0.099 ; J1 = 0.036 ; J2 = 0.0182 ;

% M
M11 = J1+Lc1^2*m1+J2+m2*Lc2^2+2*m2*Lc2*l1*cos(ap2)+m2*l1^2 ;
M12 = J2+m2*Lc2^2+m2*Lc2*l1*cos(ap2) ;
M21 = M12 ;
M22 = J2+m2*Lc2^2 ;

% Gravity
G1 = m2*g*sin(ap1+ap2)*Lc2+m2*g*sin(ap1)*l1+m1*g*sin(ap1)*Lc1 ;
G2 = m2*g*sin(ap1+ap2)*Lc2 ;

% Non-Linear Terms
C11 = -2*m2*Lc2*l1*sin(ap2)*av2 ;
C12 = -m2*Lc2*l1*sin(ap2)*av2 ;
C21 = m2*Lc2*l1*sin(ap2)*av1 ;
C22 = 0 ;
```

% PID

kp1=600; kv1=60; ki1=200;

kp2=600; kv2=60; ki2=200;

s1 =da1+ kv1*ev1 + kp1*ep1 + ki1*E1 ;

s2 =da2+ kv2*ev2 + kp2*ep2 + ki2*E2 ;

% Computed Torques

t11 = M11*s1 + M12*s2 + C11*av1 +C12*av2 +G1 ;

t22 = M21*s1 + M22*s2 + C21*av1 + G2 ;

%Disturbance

td1 = 0.1*t11 ;

td2 = 0.1*t22 ;

%Joint Torques

t1 = t11+td1 ;

t2 = t22+td2 ;

Orthesis Dynamic Model 2 Equations

function [t1,t2]= fcn(da1,da2,ap1, ap2, av1,av2,dp1,dp2,dv1,dv2,E1,E2)

ep1 = dp1-ap1;

ep2 = dp2-ap2;

ev1 = dv1-av1;

```

ev2 = dv2-av2;

m1 = 2.4 ; m2 = 2.12 ; l1 = 0.32 ; l2 = 0.26 ; g = 9.8 ;

% M
M11=(m1+m2)*l1^2+m2*l2^2+2*m2*l1*l2*cos(ap2);
M12=m2*l2^2+m2*l1*l2*cos(ap2);
M21=M12;
M22=m2*l2^2;

% Gravity
N1g = (m1+m2)*g*l1*sin(ap1)+m2*g*l2*sin(ap1+ap2);
N2g = m2*g*l2*sin(ap1+ap2);

% Non-Linear Terms
N1 = N1g + -m2*l1*l2*(2*av1*av2+av2^2)*sin(ap2);
N2 = N2g + m2*l1*l2*av1^2*sin(ap2);

% PID
kp1=600; kv1=60; ki1=200;
kp2=600; kv2=60; ki2=200;

s1 =da1+ kv1*ev1 + kp1*ep1 + ki1*E1;
s2 =da2+ kv2*ev2 + kp2*ep2 + ki2*E2;

```

% Computed Torques

$$t_{11} = M_{11}s_1 + M_{12}s_2 + N_1;$$

$$t_{22} = M_{21}s_1 + M_{22}s_2 + N_2;$$

%Disturbance

$$td_1 = 0.1*t_{11} ;$$

$$td_2 = 0.1*t_{22} ;$$

%Joint Torques

$$t_1 = t_{11} + td_1 ;$$

$$t_2 = t_{22} + td_2 ;$$

HAMM Dynamic Model 1 Equations

function [t3,t4]= fcn(da3,da4,ap3, ap4, av3,av4,dp3,dp4,dv3,dv4,E3,E4)

$$ep_3 = dp_3 - ap_3;$$

$$ep_4 = dp_4 - ap_4;$$

$$ev_3 = dv_3 - av_3;$$

$$ev_4 = dv_4 - av_4;$$

$$m_1 = 9.09 ; m_2 = 5.12 ; l_1 = 0.32 ; l_2 = 0.26 ; g = 9.8 ; Lc_1 = 0.135 ; Lc_2 = 0.118 ; J_1 = 0.1158 ; J_2 = 0.04 ;$$

% M

$$M_{211} = J_1 + Lc_1^2 * m_1 + J_2 + m_2 * Lc_2^2 + 2 * m_2 * Lc_2 * l_1 * \cos(ap_4) + m_2 * l_1^2 ;$$

$$M_{212} = J_2 + m_2 * Lc_2^2 + m_2 * Lc_2 * l_1 * \cos(ap_4) ;$$

M221 = M212 ;

M222 = J2+m2*Lc2^2 ;

% Gravity

G21 = m2*g*sin(ap3+ap4)*Lc2+m2*g*sin(ap3)*l1+m1*g*sin(ap3)*Lc1 ;

G22 = m2*g*sin(ap3+ap4)*Lc2 ;

% Non-Linear Terms

C211 = -2*m2*Lc2*l1*sin(ap4)*av4 ;

C212 = -m2*Lc2*l1*sin(ap4)*av4 ;

C221 = m2*Lc2*l1*sin(ap4)*av3 ;

C222 = 0 ;

% PID

kp1=600; kv1=60; ki1=200;

kp2=600; kv2=60; ki2=200;

s21 =da3+ kv1*ev3 + kp1*ep3 + ki1*E3 ;

s22 =da4+ kv2*ev4 + kp2*ep4 + ki2*E4 ;

% Computed Torques

t33 = M211*s21 + M212*s22 + C211*av3 +C212*av4 +G21 ;

t44 = M221*s21 + M222*s22 + C221*av3 + G22 ;

%Disturbance

td3 = 0.1*t33 ;

td4 = 0.1*t44 ;

%Joint Torques

t3 = t33 +td3 ;

t4 = t44 +td4 ;

HAMM Dynamic Model 2 Equations

function [t3,t4]= fcn(da3,da4,ap3, ap4, av3,av4,dp3,dp4,dv3,dv4,E3,E4)

ep3 = dp3-ap3;

ep4 = dp4-ap4;

ev3 = dv3-av3;

ev4 = dv4-av4;

m1 = 9.09 ; m2 = 5.12 ; l1 = 0.32 ; l2 = 0.26 ; g = 9.8 ;

% M

DM11=(m1+m2)*l1^2+m2*l2^2+2*m2*l1*l2*cos(ap4);

DM12=m2*l2^2+m2*l1*l2*cos(ap4);

DM21=DM12;

DM22=m2*l2^2;

% Gravity

$N_{11g} = (m_1 + m_2) * g * l_1 * \sin(\alpha_3) + m_2 * g * l_2 * \sin(\alpha_3 + \alpha_4);$

$N_{22g} = m_2 * g * l_2 * \sin(\alpha_3 + \alpha_4);$

% Non-Linear Terms

$N_{11} = N_{11g} + -m_2 * l_1 * l_2 * (2 * \dot{\alpha}_3 * \dot{\alpha}_4 + \dot{\alpha}_4^2) * \sin(\alpha_4);$

$N_{22} = N_{22g} + m_2 * l_1 * l_2 * \dot{\alpha}_3^2 * \sin(\alpha_4);$

% PID

$k_{p1} = 600; k_{v1} = 60; k_{i1} = 200;$

$k_{p2} = 600; k_{v2} = 60; k_{i2} = 200;$

$s_{11} = \ddot{\alpha}_3 + k_{v1} * \dot{e}_3 + k_{p1} * e_3 + k_{i1} * E_3;$

$s_{22} = \ddot{\alpha}_4 + k_{v2} * \dot{e}_4 + k_{p2} * e_4 + k_{i2} * E_4;$

% Computed Torques

$t_{33} = D_{M11} * s_{11} + D_{M12} * s_{22} + N_{11};$

$t_{44} = D_{M21} * s_{11} + D_{M22} * s_{22} + N_{22};$

%Disturbance

$t_{d3} = 0.1 * t_{33};$

$t_{d4} = 0.1 * t_{44};$

%Joint Torques

t3 = t33 +td3 ;

t4 = t44 +td4 ;