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Date : ……………………………………………………….

EMG AND CUBIC POLYNOMIAL BASED LOWER LIMB REHABILITATION DEVICE FOR POST STROKE PATIENT

TAN JIN SIANG

A report submitted in partial fulfillment of the requirements for the degree of Bachelor of Mechatronics Engineering

Faculty of Electrical Engineering

UNIVERSITI TEKNIKAL MALAYSIA MELAKA

2015

I declare that this report entitle "EMG and Cubic Polynomial Based Lower Limb Rehabilitation Device for Post Stroke Patient" is the result of my own research except as cited in the reference. The report has not been accepted for any degree and is not concurrently submitted in candidature of any other degree.

Signature : …………………………………………………..

Name : …………………………………………………..

Date : …………………………………………………..

To my beloved mother and father

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First, I want to thank to UTeM for giving me this chance to have my Final Year Project in fulfilling the Bachelor of Mechatronics Engineering and also support me in terms of money as well as the facilities in the university. Secondly, I want to thanks to my supervisor, Pn Norafizah binti Abas and Dr. Muhammad Fahmi bin Miskon for giving me the chance of becoming her student that supervise under her. She also provide me a good title and fully guide me all the way throughout this Final Year Project. Then, I want to thank to my parents and fellow friends for supporting and helping me throughout this Final Year Project.

ABSTRACT

Nowadays, stroke is the second disease killer in Malaysia. Post stroke patient are having the problem of hemiparesis and unable to walk without assistance. The lower limb rehabilitation device in the market using the different type of controller and sensor which can affect the repeatability of the system and also the accuracy of EMG sensor which is self-made is unknown. Therefore, a cubic polynomial and EMG trajectory planning lower limb rehabilitation device which is a better design compare to the device in the market is designed. The main objective of this project is to design a cubic polynomial and EMG based trajectory planning lower limb rehabilitation device for the post stroke patient to regain their walking ability. The design of lower limb rehabilitation is wearable type with 2 DOF with the training covers the thigh and knee. This is because the user is assumed to have muscle weaken at thigh and knee only but the ankle can move freely like the healthy person. The EMG sensor is self-made by using amplifier to condition the signal and using AgCl electrode to receive the EMG signal. The controller used is EMG controller which map the EMG signal with Initial Contact and Loading Response of the walking gait cycle act as an on off switch of the device. Three experiments are conducted which are data collecting for EMG value, accuracy analysis for EMG sensor and repeatability analysis for the device. The data of the EMG signal are recorded from 5 persons so that the device can be rotate based on the EMG signal. The repeatability analyze of the device is by repeating 10 times of the experiment and comparing the desired trajectory angle value with measured angle value. The accuracy of the EMG signal conditioning is analyzed by comparing the EMG signal conditioning value with muscle sensor value. After the experiments are carried out, it is found that the average EMG signal value is 0.512V for maximum and -0.2856V for minimum. The experimental results of the device of RMS error are 0.2699 radian for hip joint and 0.4037 radian for knee joint and the repeatability in terms of consistency are between the range of 0.0156 radian and 0.0612 radian for both if the joint. The accuracy of the EMG sensor is 96.86%.

ABSTRAK

Pada masa kini, strok adalah pembunuh penyakit kedua di Malaysia. Pesakit strok ischaemic selepas mengalami masalah hemiparesis dan tidak mampu berjalan tanpa bantuan. Peranti pemulihan kaki di pasaran menggunakan jenis yang berbeza daripada pengawal dan sensor yang boleh memberi kesan kepada kebolehulangan sistem. Oleh itu, merancang trajektori polinomial dan EMG padu peranti pemulihan kaki yang reka bentuk yang lebih baik berbanding dengan peranti ini di pasaran direka. Objektif utama projek ini adalah untuk mereka bentuk polinomial kubik dan EMG trajektori berdasarkan peranti pemulihan kaki untuk pesakit pasca strok mendapat semula keupayaan berjalan kaki. Reka bentuk pemulihan kaki adalah jenis boleh pakai dengan 2 DOF dengan latihan yang meliputi paha dan lutut. Ini kerana pengguna dianggap mempunyai otot lemah di paha dan lutut sahaja tetapi pergelangan kaki boleh bergerak bebas seperti orang yang sihat. Sensor EMG adalah buatan sendiri dengan menggunakan penguat kepada keadaan isyarat dan menggunakan AgCl elektrod untuk menerima isyarat EMG itu Pengawal digunakan adalah EMG pengawal yang memetakan isyarat EMG dengan Hubungi Awal dan Loading Response kitaran perbuatan gaya berjalan berjalan kaki sebagai suis off peranti. Tiga eksperimen dijalankan iaitu pengumpulan data untuk nilai EMG , analisis ketepatan sensor EMG dan analisis kebolehulangan untuk peranti. Data isyarat EMG yang direkodkan dari 5 orang supaya peranti boleh putar berdasarkan isyarat EMG itu. The kebolehulangan menganalisis peranti ini dengan mengulangi 10 kali percubaan dan membandingkan nilai sudut trajektori dengan nilai sudut diukur. Ketepatan penyesuaian isyarat EMG dianalisis dengan membandingkan nilai penyesuaian isyarat EMG dengan nilai sensor otot. Selepas eksperimen dijalankan, didapati bahawa purata nilai isyarat EMG adalah 0.512V untuk maksimum dan -0.2856V untuk minimum. Keputusan eksperimen peranti ralat RMS adalah 0.2699 radian bagi sendi pinggul dan 0.4037 radian untuk sendi lutut dan konsisten adalah antara pelbagai 0.0156 dan 0.0612 radian radian untuk kedua-dua jika sendi. Ketepatan sensor EMG adalah 96.86 %.

TABLE OF CONTENT

LIST OF TABLES

LIST OF FIGURES

LIST OF SYMBOL

θ(t)	-	Angle
$\frac{\delta\theta_{i-1}}{\delta t_{i-1}}$	-	Angular velocity
a_0, a_2, a_3	-	Unknown
R	-	Resistor
f	-	Frequency
C	-	Capacitor

LIST OF APPENDICES

Chapter 1

INTRODUCTION

1.1 Research Background

According to World Health Organization (WHO) [1], stroke is caused by the blood vessel burst due to the clot blocking the blood vessel. This will cut off the supply of the oxygen, nutrients and causing the brain damage. According to The Star: Health [2], there are 2 major type of stroke which are ischaemic and haemorrhagic. Ischaemic stroke occurs when a blood vessel in the brain are blocked. Haemorrhagic stroke occur when a blood vessel inside the brain is weak or burst and causing the blood leak into the brain. The risk of having stroke increase from age to age as shown in the Table 2. From The Star: Health [2], the people after the age of 55 will having the double increase in the risk of having stroke for every decade the person is alive. There are also some other risk that can increase the risk of having stroke like hypertension, diabetes, atrial fibrillation and so on. According to My Malaysia Health [3], stroke is second disease killer in Malaysia as shown in Table 1.1.

Table 1.1: Top disease killer in Malaysia (source: My Malaysia Health) [3]

No	Top 5 Cause of Death	%
1	Coronary heart disease	22.18
2	Stroke	11.67
3	Influenza & pneumonia	9.2
4	Road traffic accidents	7.85
5	HIV/ AIDS	5.53

According to National Stroke Association [4], about 50% of the acute ischemic post stroke patient having a problem which is called hemiparesis. Hemiparesis is the weakness of the muscle of one sided of the body. This will cause the disability of walking, grabbing things and also eating due to the muscle weaken of the one sided of the body and the patient usually will have more than one disability problem.

1.2 Motivation

According to Genentech USA, Inc [5], about 50% of acute ischemic stroke which is one type of stoke will having hemiparesis and 30% of the patient unable to walk without assistance which is the top 1 and top 3 of the disabilities as shown in Figure 1.1. Difficulty of general movement like walking and stabilizing are due to muscle weaken which also called impaired motor control. Impaired motor control of one sided body is called hemiparesis. This disability is the most factor that nowadays most of the post stroke patient are having this disability. Most of the patients will have more than one kind of disability.

The stroke disease in Malaysia is also very serious. It is a second disease killer in Malaysia which is 11.67% as shown in Table 1.1. This means that there are many people in Malaysia having this kind of disease. The post stroke patient in Malaysia also have the disabilities like hemiparesis and unable to walk without assistance.

Nowadays, lower limb rehabilitation device is an important and the only device that can help the post stroke patient who have hemiparesis problem and unable to walk without assistance to regain their walking ability by having training at their lower limb. Therefore, lower limb rehabilitation training for the post stroke patient is needed to allow them to regain the skills and having a normal life like before the stroke.

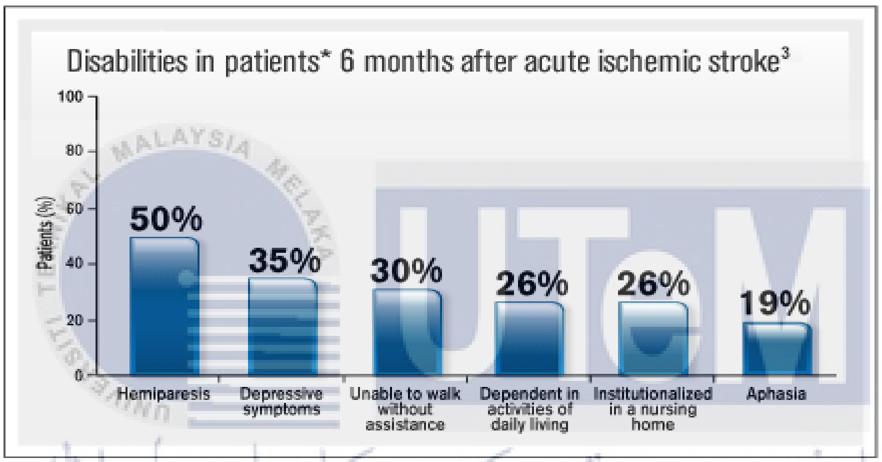

Figure 1.1: Stroke patient disability statistic (source: Genentech USA, Inc) [5]

1.3 Problem Statement

Based on Figure 1.1, hemiparesis and unable to walk without assistance are the most common disabilities faced by post stroke patients who had experienced at least 6 months of acute ischemic. Both disabilities limits or impairs patient's ability to walk which affect their quality of life and it demands resources for assistive and rehabilitation. Thus, rehabilitation device for lower limb is considered as crucial to restore the walking ability of the post stroke patients. Therefore, rehabilitation device for lower limb is a must for post stroke patient who have this kind of disability to restore their walking ability.

From Iñaki Díaz [6], the lower limb rehabilitation system can be categorized into two types which are wearable and non-wearable. The wearable which is used in both environment, indoor and outdoor, while the non-wearable can be used in indoor only. The non-wearable usually need a treadmill to work to and a therapist as training assistant. It also only allowed the patient to walk in predetermined path. Hence, the wearable type of rehabilitation device is preferred. Even though it might offer complexity in fundamental design, but it have more benefits than the non-wearable rehabilitation system. The wearable rehabilitation system can allow the user to walk their own path rather than just the predetermined path which is very noticeable that almost all systems reviewed have been commercialized. Besides that, the wearable rehabilitation system does not need the treadmill to work. However, rehabilitation training for a wearable rehabilitation system in the market mostly only cover one part of the lower limb only or knee and ankle only or hips, knee and ankle but the training for the hips, knee and ankle is costly due to an extra actuator needed. Moreover, there are don't have the training for the hips and knee. Therefore, a hips and knee wearable training rehabilitation system is needed for the patient having disability at hips and knee.

According to Iñaki Díaz [6], there are so many type of controller in the lower limb rehabilitation system in the market like Fuzzy logic, PID and so on. Most of the controller are high performance. However, those controllers involves complex algorithm and calculation. Therefore, a simple but with high performance controller is needed for the lower limb wearable rehabilitation system. Therefore to reduce the complexity of the controller, the EMG controller can be implemented in the system as a primary controller to control the lower limb rehabilitation system.

Different type of device using different type of trajectory planning. Different type of trajectory planning could cause the different effect. By using EMG as the controller could cause the consistency of the device decrease but can increase in terms of time response. The cubic polynomial could cause the response time of the device decrease but it will increase the consistency and accuracy of the device. As a summary, a cubic polynomial and EMG based trajectory planning lower limb wearable rehabilitation device with EMG as a primary controller which cover the hips and knee is designed in solving the problems.

1.4 Objective

1. To design a cubic polynomial and EMG based trajectory planning lower limb rehabilitation device for the post stroke patient to regain their walking ability.
2. To analyze overall system performance in terms of accuracy and repeatability.

1.5 Scope

1. In this project, the patient is assumed to have a hemiparesis problem with weaken muscle at thigh and knee but the muscle function normal at the ankle. Therefore, the lower limb rehabilitation prototype is only cover thigh and knee.
2. The test bed used is improved based on crutches developed by previous student, Yong Xian.
3. The prototype has 2 DOF which indicates the rotational movement at hip and knee only.
4. The EMG controller only act as an on off switch to switch on or off the designed rehabilitation device.
5. The trajectory planning of the device is based on the normal forward walking gait of a healthy person.
6. The AgCl electrode is placed at the rectus femoris muscle to detect the EMG of the lower limb and the own signal conditioning is used to condition the signal of muscle.
7. The performance of overall system is analyzed in terms of accuracy and repeatability by analyzing the accuracy of controller signal receiving from the lower limb by comparing them with the muscle sensor and the repeatability prototype is analyzed by comparing the trajectory planning with angle of the device with 10 times of repetition.

1.6 Report Outline

This report is about the development of lower limb rehabilitation device for post stroke patient. In this report, the chapter 1 covers about the research background of the post stroke patient and motivation for designing a lower limb rehabilitation device for post stroke patient. The objective and scope of the design is stated in this chapter. The theory and basic principle and the review of previous related work of the lower limb rehabilitation device is covered in chapter 2 of the report. This report also lists out the methods and techniques used in this design which stated in chapter 3. The result and the conclusion are covered in chapter 4 and chapter 5 in this report.

Chapter 2

LITERATURE REVIEW

2.1 Theory and Basic Principle

Theory and basic principle of the lower limbs rehabilitation device for the post stroke patient is discussed in details in this section. It covers stability of human body, lower limb muscle, electromyography, post stroke patient, weight bearing for the lower limb as well as its motion during walking.

2.1.1 Basic Principle of Lower Limb Rehabilitation Device

2.1.1.1 Post Stroke Patient

In this section, the research will discusses about the disability of the post stroke patients. According to National Institute of Neurological Disorder and Stroke [10], the post stroke patients will have several disability in physical and also have some problem in their mental. The problems like problems of controlling movement, problems of using or understanding the language, sensory disturbance and also emotional disturbance.

According to National Stroke Association [4], about 80% of the post stroke patients will have a problem which is called hemiparesis. Hemiparesis is the one sided muscle weakness especially at the arm and lower limb. This can affect the motion of the arm and also the lower limb like walking or grabbing things.

From National Stroke Association [4], hemiparesis can be separated into 2 parts which are right sided and left sided. The right sided hemiparesis will have the problem of using or understanding the language. The left sided hemiparesis will have memory problems. Damage of the lower brain will cause the motion failure of the arm and also the lower limb which is called ataxia.

2.1.1.2 Motion of the Lower Limb during Walking

According to Aaron M. Dollar [13], there are 8 stages of the motion while a human walking. Figure 4 shows the one gait cycle when a human walking. There are 2 important phase for the gait cycle which is stance phase and swing phase. The stance phase is the phase when the human body moves forward and the swing phase is when the lower limb back into the initial position.

From Aaron M. Dollar [13], human lower limb contains of 7 degree of freedom (DOF) which is the rotation of the ankle, abduction-adduction of the ankle, flexion-extension of the ankle, flexion-extension of the knee, rotation of the thigh, abduction-adduction of the thigh and flexion-extension of the thigh. For a human to walk forward, the lower limb only uses 3 DOF which is flexion-extension of the thigh, flexion-extension of the knee and flexion-extension of the ankle.

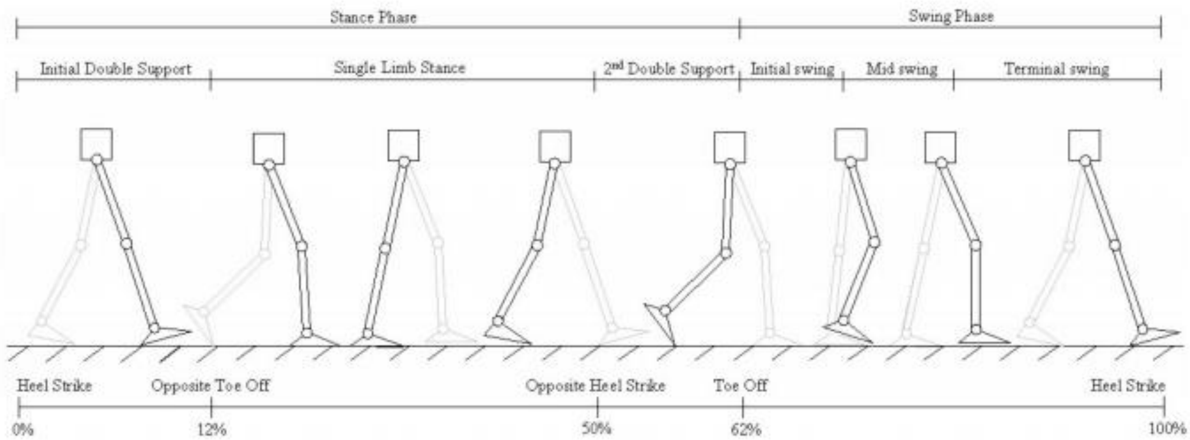

Figure 2.1: Motion of the lower limb during walking (source: Lower Extremity Exoskeletons and Active Orthoses: Challenges and State-of-the-Art) [13]

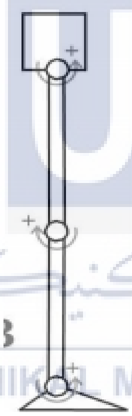

Figure 2.2: DOF of the lower limb (source: Lower Extremity Exoskeletons and Active Orthoses: Challenges and State-of-the-Art) [13]

2.1.1.3 Center of Gravity and Stability

Based on the book, Scientific Basis of Human Motion [7], the center of the gravity of human body is different on height, body size and also as well as the gender. The standing position will affect the stability of the human body. Female center of gravity is 55% of the body height and male is 57% of the body height.

According to Nancy Hamilton et. al. [7], the stability of the human body can be increased with the help of some instrument like using a crutch while walking or standing. The crutch can increases the base support area of the human body. With the increase of the base support area, the stability of the human body will also increases.

Figure 2.3: Center of gravity of human body (source: *Kinesiology : Scientific Basis of Human Motion)* [7]

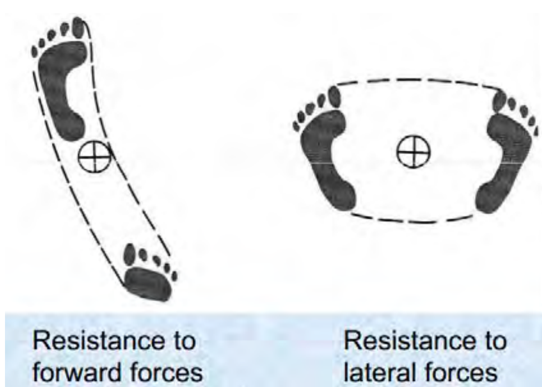

Figure 2.4: Shape of the base support of the human body based on the standing position (source: *Kinesiology : Scientific Basis of Human Motion)* [7]

2.1.1.4 Weight Bearing of the Lower Limb

The average of the human lower limb is 8kg for the person that weight 51kg. For the hemiparesis patient, according to Ruth Dickstein [11], the hindfoot can stands for 39% of the body weight of the post stroke patient and 50% for forefoot before the rehabilitation of the lower limb treatment. After the treatment, the hind foot can stands for 47% of the body weight which increase 8% but the forefoot still remain the same which is 50% of the body weight.

2.1.1.5 Human Height

According to Plagenhoef, S., Evans [12], percentage of the length of thigh is 24.05% of the overall body height and knee length is 25.2% of the overall body height. This means that for group of persons who are about 160cm to 180cm, their length of thigh and leg are 38cm to 43cm and 40cm to 45cm.

2.1.2 Basic Principle of EMG Sensor

2.1.2.1 Electromyography (EMG)

According to Konrad, P [9], EMG is a myoelectric signal which produced by the muscle fiber membranes. The EMG in nowadays is widely used in many fields especially in rehabilitation for training purpose. It can be detected by using the muscle sensor.

According to Konrad, P [9], there are few types of electromyography like Surface Electromyography, Fine Wire Electromyography, Neuromuscular Electrical stimulation and EMG- Triggered stimulation. Most of the application is using the Surface Electromyography which detect the muscle signal by attaching the sensor on the surface of the muscle due to its easy to use and easy to get the signal from the muscle.

According to Konrad, P [9], the EMG signal processing contains of three parts which are signal conditioning, normalization and classification. The signal conditioning consists of rectifier, amplifier, bandpass filter and smoothing the signal. The normalization consists of amplitude normalization, time normalization and related parameters. The classification is the on or off of the muscle of the human body either in lower limb or upper limb.

From Konrad, P [9], the accuracy of the EMG is tested by comparing the shape of the graph of the rectified EMG signal with the EMG controller prototype. The EMG controller is tested using the same muscle and performing the same activities during the test and it also needs to depend on which part of the body need to be tested. The EMG controller needs to be test for a number of times to ensure the accuracy of the EMG when comparing with the rectified EMG signal.

2.1.2.2 Lower Limb Muscle

From He. H. et. al. [8], the lower limb comprises of many muscles that assist the body movement as well as daily activities that involve the lower limb motion. Flexion is the motion of the leg that moves forward and extension is the motion of the leg that moves backward. Adduction is the motion of the leg that moves outwards and abduction is the motion of the leg that moves inwards.

There are some of the muscles that help the motion of the thigh and leg:

- Iliacus – for flexion of the thigh.
- Psoas major – for flexion of the thigh
- Tensor fasciae latae – flexion and abduction of the thigh

- Adductor magnus – for flexion and adduction of the thigh
- Gluteus maximus – for extension of the thigh
- Rectus femoris – for leg extension
- Vastus lateralis – for leg extension
- Semitendinosus – for leg flexion
- Semimembranosus – for leg flexion

Figure 2.5: Muscle of the lower limb (source: A Study on Lower-Limb Muscle Activities during Daily Lower-Limb Motions) [8]

2.2 Review of Previous of Related Works

In this section, the research is about the lower limb robotics rehabilitation system. This section will discuss about the types of the robotic rehabilitation system for the lower limb that existing in the market and also the few design of lower limb rehabilitation system.

2.2.1 Types of Lower Limb Robotics Rehabilitation System

According to Iñaki Díaz [6], nowadays, there are many kind of lower limb robotics rehabilitation system in the market. There are 2 types of the rehabilitation system which is non-wearable types as shown in Figure 2.6 and wearable types as shown in Figure 2.7. The non-wearable type rehabilitation system that contain a multifunction big machine which record the reading of the training and improve the training but need the therapist to assist and required a treadmill as shown in the Figure 2.6. The wearable type rehabilitation system is the system can use at the outdoor and also can use it without the help of a therapist but only contain one function. However, most of the wearable type rehabilitation system is still in trial stage and most of them is for the knee and ankle training.

According to Bortole, M. [14], the type of the design is wearable lower limb rehabilitation system which is covered thigh, knee and ankle and used 6 DOF as shown in Figure 2.8. The lower limb rehabilitation system design by Enguo, C. [15] is non-wearable lower limb rehabilitation system which is only covered thigh and knee and used 2 DOF as shown in Figure 2.9. Same as Enguo, C, the lower limb rehabilitation system design by Koceska, N. [16] is non-wearable lower limb rehabilitation system but cover thigh, knee and ankle and it uses 10 DOF as shown in Figure 2.10. The Figure 2.11 shows the design which is a non-wearable lower limb rehabilitation system designed by Kari A. Danek [17]. The design only covers the ankle and it is 1 DOF.

According to Erhan Akdogˇan [18] as shown in the 2.12, the lower limb rehabilitation system is a non-wearable and 3 DOF device which covers thigh and knee only. The lower limb rehabilitation system design by Yixiong Chen [19] is non-wearable but it covers thigh, knee and ankle and it is a 3 DOF device as shown in the Figure 2.13. Figure 2.14 (Nelson Costa [20]) shows the design which is a wearable and 10 DOF lower limb rehabilitation system. It covers thigh, knee and ankle. Figure 2.15 (Carberry et al [21]) and Figure 2.16 (Park et al. [21]) shows the designs which are wearable and they are 2 DOF lower limb rehabilitation device that cover ankle only. Active modular elastomer for soft wearable assistance robots (Park et al. [21]) as shown in the Figure 2.17 is also a 2 DOF and wearable lower limb rehabilitation device but covers knee only.

Figure 2.6: Non-wearable type rehabilitation system [7]

Figure 2.7: Wearable type rehabilitation system [7]

Figure 2.8: Lower limb rehabilitation system designed by Bortole, M. [14]

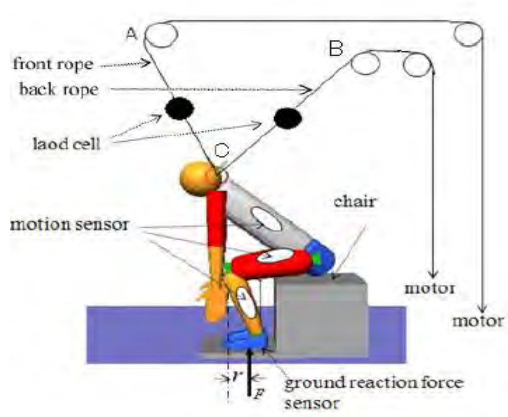

Figure 2.9: Lower limb rehabilitation system designed by Enguo, C [15]

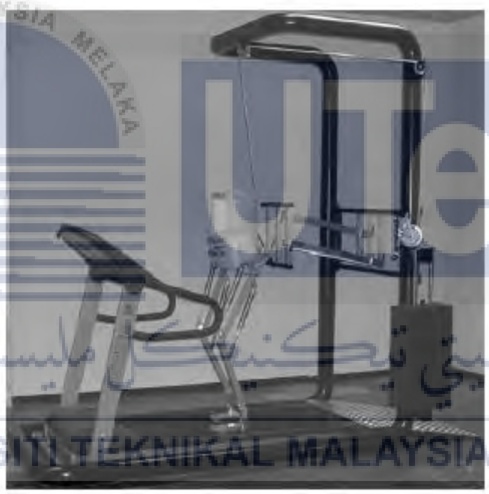

Figure 2.10: Lower limb rehabilitation system designed by Koceska, N [16]

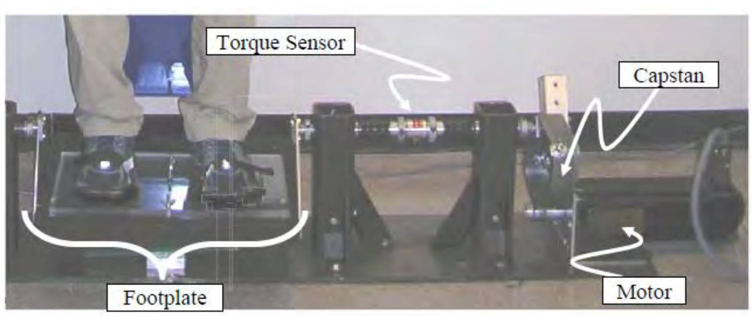

Figure 2.11: Lower limb rehabilitation system designed by Kari A. Danek [17]

Figure 2.12: Lower limb rehabilitation system designed by Erhan Akdog˘an [18]

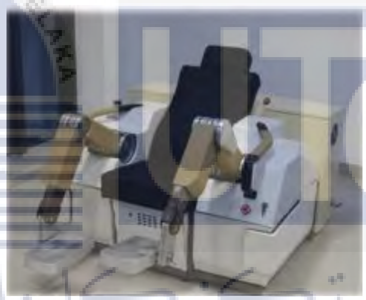

Figure 2.13: Lower limb rehabilitation system designed by Yixiong Chen [19]

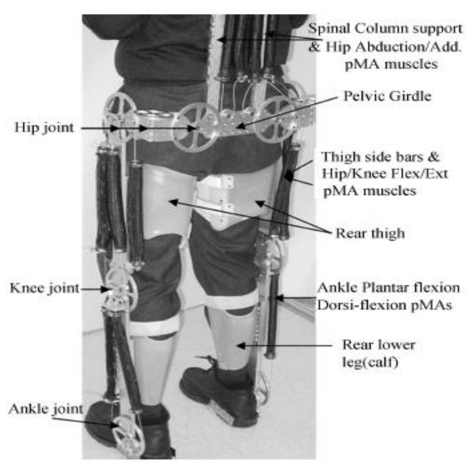

Figure 2.14: Lower limb rehabilitation system designed by Nelson Costa [20]

Figure 2.15: Active ankle-foot orthosis (AAFO) designed by Carberry et al. (source: Recent Trends in Lower-Limb Robotic Rehabilitation Orthosis: Control Scheme and Strategy for Pneumatic Muscle Actuated Gait Trainers) [21]

Figure 2.16: Bio-inspired active soft orthotic device for ankle foot pathology designed by Park et al. (source: Recent Trends in Lower-Limb Robotic Rehabilitation Orthosis: Control Scheme and Strategy for Pneumatic Muscle Actuated Gait Trainers) [21]

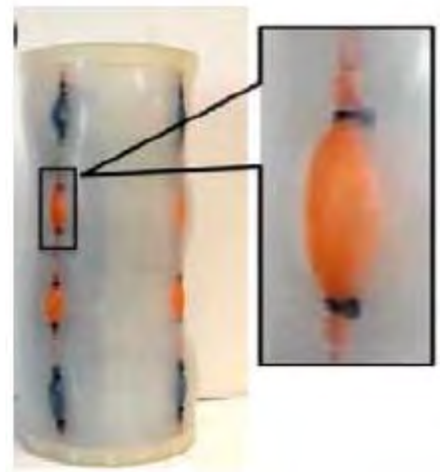

Figure 2.17: Active modular elastomer for soft wearable assistance robots designed by Park et al. (source: Recent Trends in Lower-Limb Robotic Rehabilitation Orthosis: Control Scheme and Strategy for Pneumatic Muscle Actuated Gait Trainers) [21]

2.2.2 Trajectory Planning

According to Magdo Bortole [14] and Nelson Costa [20], only both of them using the cubic polynomial as the trajectory planning while Enguo, C. [15], Koceska, N. [16], Yixiong Chen [19] and Erhan Akdogˇan [18] are using the inverse kinematics as the trajectory planning. The cubic polynomial trajectory planning in their design are used to control the movement of lower limb rehabilitation device based on the desired angle either sit and stand angle or walking motion angle. However, the inverse kinematics trajectory planning in their design are used to control the movement of lower limb rehabilitation device based on the force of the lower limb to control the torque of the actuator.

2.2.3 Controller

In Enguo, C. [15] design, the controller used is EMG which using the EMG signal and convert it into forces to calculate the muscle forces and the position of the actuator will be controlled based on the muscle forces. According to Bortole, M. [14], Koceska, N. [16], Kari A. Danek [17] and Nelson Costa [20], the designs are using the Fuzzy Logic and PID as the controller to control the actuator and also the position of the lower limb rehabilitation system. Both of them involve many calculation. In Erhan Akdogˇan [18] and Yixiong Chen [19], they are using the impedance as the controller to control the lower limb degree of movement. This type of controller has algorithm that contain calculation. Active ankle-foot orthosis (AAFO) designed by Carberry et al. [21] using the feedback control that utilizes a fuzzy logic gait phase detection system as the controller while bio-inspired active soft orthotic device for ankle foot pathology designed by Park et al. [21] using the feed-forward and feedback controllers as the controller of the device. Both of this controller is used to control the ankle joint angle. Active modular elastomer for soft wearable assistance robots designed by Park et al. [21] control the motion of the knee through shape and rigidity control.

2.2.4 Sensor

In Enguo, C. [15] design, he use the motion sensor to sense the motion of the rehabilitation system and hard type sensor to receive the signal from the lower limb muscle. The main sensor used in Bortole, M. [14], Koceska, N. [16] and Nelson Costa [20] are potentiometer to detect the motion of the device. According to Kari A. Danek [17], Erhan Akdogˇan[18] and Yixiong Chen [19] designs, they use the torque sensor as their main sensor to detect the force between the lower limb and the device. Active ankle-foot orthosis (AAFO) designed by Carberry et al. [21] using the force sensory resistor to identify contact between foot and ground and rotary encoder to measure the angular displacement of ankle joint. Bio-inspired active soft orthotic device for ankle foot pathology designed by Park et al. [21] and active modular elastomer for soft wearable assistance robots designed by Park et al. [21] using the strain sensor as the primary sensor to measure the angle of ankle joint.

2.2.5 Performance Analysis from Previous Related Work

According to Bortole, M. [14], the performance analysis of this design is using a high speed infrared camera. The lower limb is attached to a marker that can reflect the infrared. This test can used to detect the angle of the lower limb. In Enguo, C. [15] design, the performance of the design was analyzed by using a video camera in measuring the movement of the lower limb. This method is using the repeatability test to test the performance by using the design testing on few person. Based on Koceska, N. [16] design, this design using a video camera, Panasonic Nv-gs400eg with 4.0 MPixel, 3CCD, the Leica Dicomar objective and a focal distance of 3.3mm to analyze the repeatability of the design. The camera is set 13m from the treadmill and the lower limb is marked with three marker. Then the distance of the lower limb movement is measured using the pixel of the camera by transforming it into measuring units. The results from the experiments, at a confidence level $p=0.05$, have shown that the maximum absolute error is 3.5mm, which has no significant impact on the rehabilitation process and can be treated as an acceptable result. According to John J. Craig [22], the design can be analyzed by comparing the trajectory planning with the angle of the robot. According to Sharif Muhammad Taslim Reza [23], the EMG signal is perpendicular to the angle of the lower limb movement. Therefore, the EMG signal can be compare with the angle and use the value for programming the device.

2.3 Summary

Based on the Table 2.1, there are 7 example of design. There are 5 non-wearable and 5 wearable type of lower limb rehabilitation system. According to the Bortole, M. [14] and Nelson Costa [20] designs, their designs are wearable which are more preferable from the user compare to the non-wearable designs (Iñaki Díaz [6]). There are 4 designs are using the inverse kinematics as the trajectory planning and 2 designs are using the cubic polynomial as the trajectory planning. The cubic polynomial trajectory planning is used to control the movement of lower limb rehabilitation device based on the desired angle either sit and stand angle or walking motion angle but the inverse kinematics trajectory planning in their design are used to control the movement of lower limb rehabilitation device based on the force of the lower limb to control the torque of the actuator. Since the post stroke patient is assume to have weaken muscle at thigh and knee, therefore it is suitable to use cubic polynomial as the trajectory planning rather than inverse kinematics because the inverse kinematics requires the patient to have force at the lower limb to move device.

Different sensor is used in each of the design. However, the performance of all the sensor used is almost the same but costly due to its high accuracy. Therefore, the design of the lower limb rehabilitation system can use the own signal conditioning as EMG sensor so that to reduce the cost but remain the performance.

The device controller is used as a switch which switch on and off the lower limb rehabilitation device. In order to have that kind of controller, a simple controller like EMG controller can be implemented into the device as a switch. Moreover, PID, Fuzzy Logic and impedance controller have more calculation and algorithm than the EMG controller which is used by Enguo, C. [15]. Therefore, in order for the controller used as an on off switch in the device, the EMG controller which do not have complex calculation and algorithm like others controller is the best choice among them.

Based on Table 2.1, there are no wearable lower limb rehabilitation system which cover thigh and knee. Therefore, a 2 DOF wearable cubic polynomial based trajectory planning lower limb rehabilitation device which covers the thigh and knee with EMG as the controller and own signal conditioning as the EMG sensor is needed for the post stroke patient to regain their walking ability.

From the research of the literature review, the best performance analyzing of the lower limb rehabilitation system design is repeatability analysis and accuracy analysis. The prototype can be analyzed either in angle of the lower limb or distance of the lower limb movement as stated in the in section 2.2.5. However, those methods need the video camera to help measuring. This is very costly due to the video camera. However, the performance of the prototype can be analyzed by measuring the angle of the lower limb rehabilitation system with trajectory planning. The EMG sensor is analyzed by comparing the peak to peak voltage value of the muscle sensor but with the condition of attached it with the same muscle and performing the same activities during the test. The value can be obtained by using the oscilloscope. The EMG sensor need to be measured for at least 3 times to ensure the accuracy of the EMG when comparing with the muscle sensor.

Table 2.1: List of detail of the lower limb rehabilitation system

Title	Author	Type	Position of lower limb	DOF	Trajectory Planning	Sensor	Controller
Design and Control of a Robotic Exoskeleton for Gait Rehabilitation	Bortole, M.	Wearable	Thight, knee and ankle	6	Cubic Polynomial	Potentiometer and strain gauge	PID and Fuzzy Logic
Human Lower Limb Dynamic Analysis Using Wearable Sensors and Its Applications to a Rehabilitation Robot	Enguo, C.	Non-wearable	Thight and knee	2	Inverse Kinematics	Motion Sensor and hard type sensor	EMG
Control Architecture of a 10 DOF Lower Limbs Exoskeleton for Gait Rehabilitation	Koceska, N.	Non-wearable	Thight, knee and ankle	10	Forward and Inverse Kinematics	Potentiometer	Fuzzy Logic
Lower Limb Motor Coordination and Rehabilitation Facilitated through Self-Assist	Kari A. Danek	Non-wearable	Ankle	1	NA	Torque sensor	PID controller

The design and control of a therapeutic exercise robot for lower limb rehabilitation: Physiotherabot	Erhan Akdog˘an	Non-wearable	Thigh and knee	3	Inverse Kinematics	Torque sensor	Impedance controller
The FES-assisted control for a lower limb rehabilitation robot: simulation and experiment	Yixiong Chen	Non-wearable	Thigh, knee and ankle	3	Inverse Kinematics	Torque and angle sensor, EMG sensor	Impedance controller
Joint Motion Control of a Powered Lower Limb Orthosis for Rehabilitation	Nelson Costa	Wearable	Thigh, knee and ankle	10	Cubic Polynomial	Potentiometer	PID controller
Recent Trends in Lower-Limb Robotic Rehabilitation Orthosis: Control Scheme and Strategy for Pneumatic Muscle Actuated Gait Trainers	Yamamoto, S Designed by: Carberry et al.	Wearable	Ankle	2	NA	Force sensory resistor and rotary encoder	Feedback control that utilizes a fuzzy logic gait phase detection system

Recent Trends in Lower-Limb Robotic Rehabilitation Orthosis: Control Scheme and Strategy for Pneumatic Muscle Actuated Gait Trainers	Yamamoto, S Designed by: Park et al.	Wearable	Ankle	2	NA	Strain sensor	Feed-forward and feedback controllers
Recent Trends in Lower-Limb Robotic Rehabilitation Orthosis: Control Scheme and Strategy for Pneumatic Muscle Actuated Gait Trainers	Yamamoto, S Designed by: Park et al.	Wearable	Knee	2	NA	Strain sensor	Through shape and rigidity control

Chapter 3

RESEARCH METHODOLOGY

This chapter is about methods and techniques which are used in this lower limb rehabilitation device to achieve the objectives. The detail of the experiment set up will also be discussed in this section.

Figure 3.1 shows the overall methodology flow chart of this research. Firstly, literature review is conducted to perform gap analysis. Based on that, motivation, problem statement and objectives of this research are conducted. After that, the lower limb rehabilitation system is designed and tested. If the device have problem, then the troubleshooting is carried out. The design of lower limb rehabilitation system is discussed in section 3.1. In this section, device design, material selection, trajectory planning and software development are included. Then, EMG signal processing which includes the schematic circuit is conducted at section 3.2. If the circuit have problem, then troubleshooting is carried out to solve the problem. Finally, overall design performance analysis which contains of the performance analysis procedure is conducted at section 3.3.

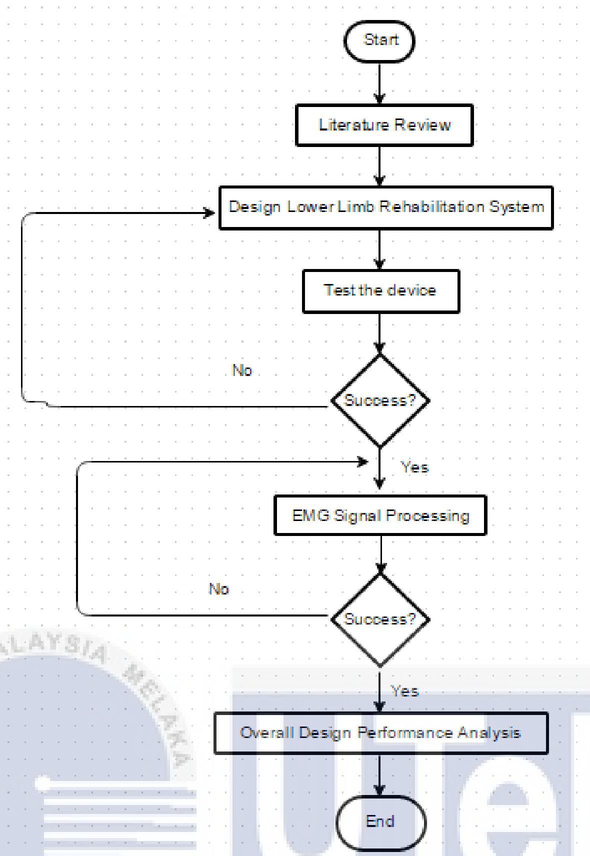

Figure 3.1 Overall methodology flow chart

3.1 Designing the Lower Limb Rehabilitation System for Post Stroke Patient

3.1.1 Overall System

Figure 3.2 shows block diagram of lower limb rehabilitation system. The EMG controller is used as the controller for mapping the EMG signal, servo motor is used as actuator for the motion of system and EMG signal conditioning is used for condition the EMG signal which receives from AgCl electrode.

Material used for the system is Aluminum Alloy due to its light weight and hard material characteristic. EMG signal conditioning is a self-made EMG sensor which will be discuss in section 3.2. The servo motor is Sonxun SXSV50 12 - 24 V High Power Servo Motor as shown in Appendices A. This motor is a type of DC geared motor which convert the DC motor to servo motor. Microcontroller used is Arduino UNO board as shown in Appendices A due to its short and easy understanding coding. This microcontroller is used as a medium for the EMG controller.

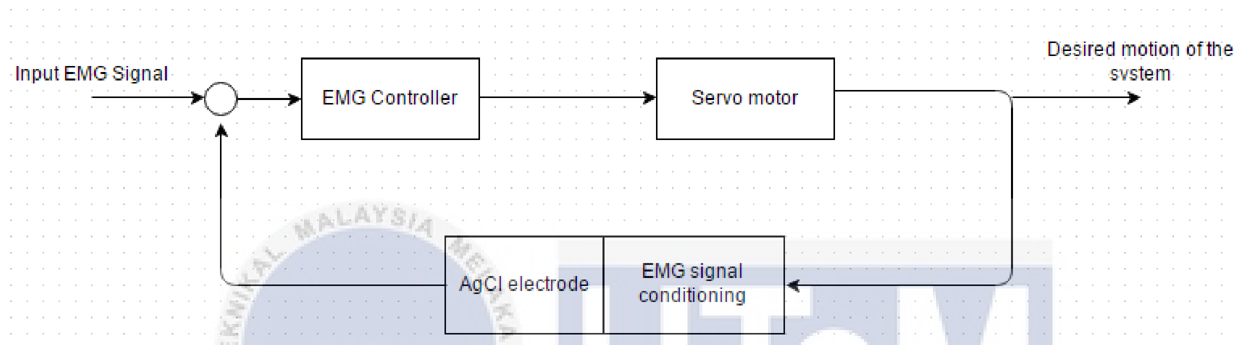

Figure 3.2: Block diagram of lower limb rehabilitation system

3.1.2 Trajectory Planning

Lower limb rehabilitation system is designed using 2 DOF which covers thigh and knee. The leg is tied with the design using ropes. Figure 3.3 shows the free body diagram of the design. The hip joint is fixed which means that when hip joint rotate, the parts below hip joint will move and hip joint will not move. Servo motors is placed at the hip and knee joint which can rotate in terms of angle.

The purpose of the trajectory planning is to find the desired angle of the device so that the device can perform with the desired angle. The desired angle is based on the degree of motion of human walking.

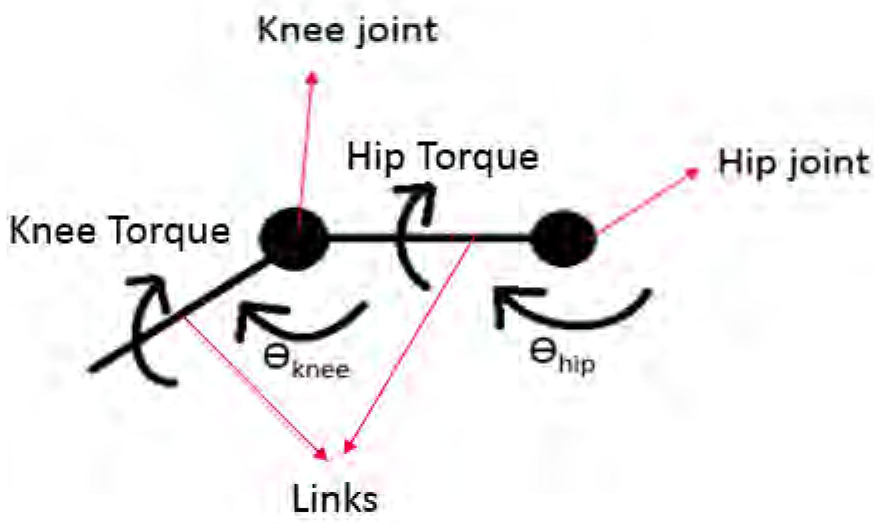

Figure 3.3: Free body diagram of the lower limb rehabilitation system

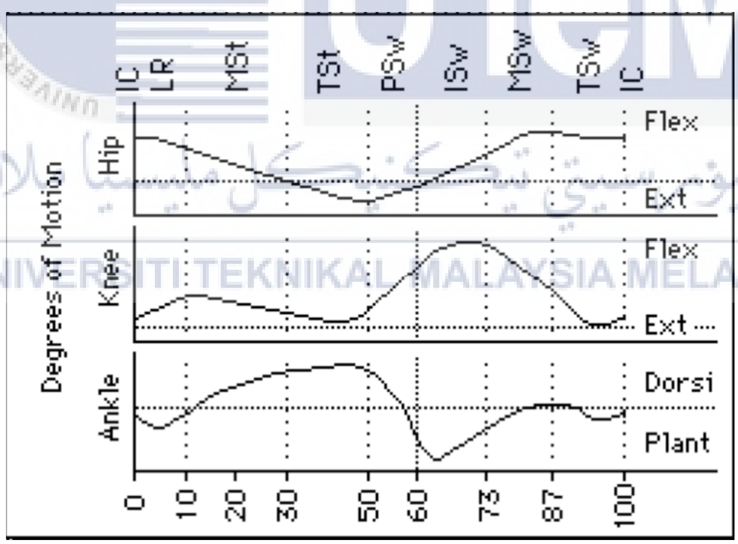

Figure 3.4: Degree of motion of the lower limb during walking (source: Rancho Los Amigos National Rehabilitation Center) [24]

Table 3.1: Degree of motion based on the hip and knee

	Degree of Motion (°)							
Joint\Phase	IC	LR	MST	TST	PSW	ISW	MSW	TSW
Hip	30	30	5	-10	0	25	35	30
Knee	5	15	8	5 then 12	40	60 then 55	20	0 then 5

According to Rancho Los Amigos National Rehabilitation Center [24], the degree of motion of human walking is shown in Figure 3.4. In this design, the hip joint and knee joint degree of motion are taken. Table 3.1 show the degree of motion of the hip and knee which will be taken as the reference to calculate the desired velocity of the device.

From John J.Craig [22],

The initial degree, θ of the slope of knee and hip joint changes sign at the via point,

$$\theta(t_{i-1}) = \theta_{i-1}, where\ i = 1,2,3, ... \tag{3.1}$$

The final degree, θ of the slope of knee and hip joint changes sign at the via point,

$$\theta(t_i) = \theta_{i,}\ where\ i = 1,2,3, ... \tag{3.2}$$

The initial and final velocity,

$$\frac{\delta\theta_{i-1}}{\delta t_{i-1}} = \frac{\delta\theta_i}{\delta t_i} = 0 \tag{3.3}$$

$$joint\ angle, \theta(t) = a_0 + a_2t^2 + a_3t^3 \tag{3.4}$$

$$angular\ velocity, \frac{\delta\theta(t)}{\delta t} = 2a_2t + 3a_3t^2 \tag{3.5}$$

$$a_0 = \theta_{i-1} \tag{3.6}$$

$$a_2 = \frac{3}{t^2}(\theta_i - \theta_{i-1}) \tag{3.7}$$

$$a_3 = -\frac{2}{t^3}(\theta_i - \theta_{i-1}) \tag{3.8}$$

3.1.3 Mechanical Design

The mechanical design consists of 2 main parts which are hip joint and knee joint which is designed by previous student, Yong Xian. The rehabilitation device that has been designed by using Solidworks comprises of few components as shown in Appendices B and Appendices C. Figure 3.5 shows a complete hardware of the device which consists of EMG circuit, Arduino UNO board, wooden block which use to hang the device, hip joint and knee joint.

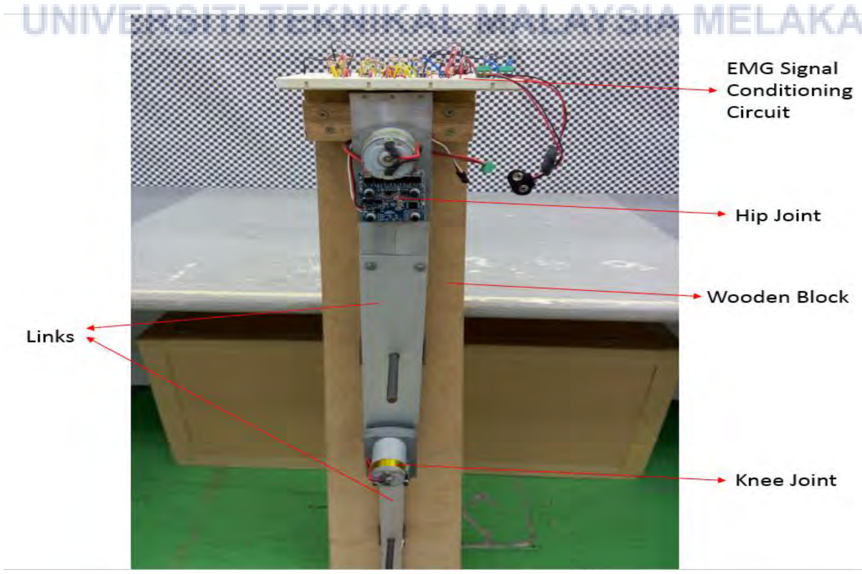

Figure 3.5: Complete Hardware of the Device

3.1.4 Bill of Component

Table 3.2 shows the quantity of components are used in the lower limb rehabilitation device. The quantity of Sonxun SXSV50 12 - 24 V High Power Servo Motor is 2 due to the motor is placed at hip and knee joint. This device also requires 1 Arduino UNO board to control the servo motor and receive the EMG signal from the EMG sensor. The EMG sensor is self-made which means that the EMG sensor no need to buy for the device.

Table 3.2 Bill of component

Component	Quantity
Sonxun SXSV50 12 - 24 V High Power Servo Motor	2
Arduino UNO board	1
EMG sensor	Self-made

3.1.5 Software Development

3.1.5.1 Software Design

Figure 3.6 shows the flow chart of the program of lower limb rehabilitation system. First, the EMG signal from the lower limb muscle need to be detected then only the servo motor rotate to the desired angle. The servo motor is rotated based on the trajectory planning. The servo motor stop after 1 cycle if there is no input EMG signal. If emergency stop button is pushed, during the servo motor rotate or before the servo motor rotate, the motor stop. Appendices D shows the program coding of the device.

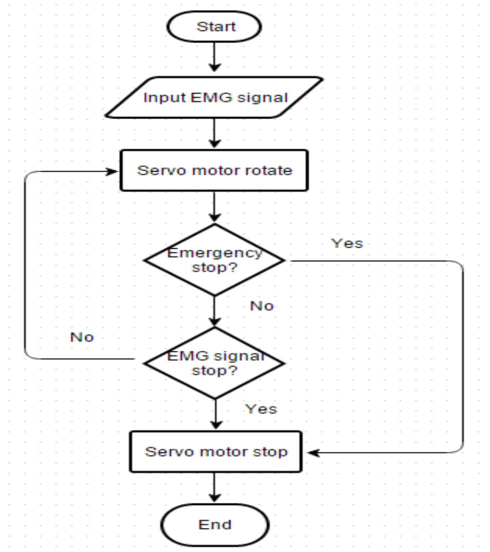

Figure 3.6: Flow chart of the program of lower limb rehabilitation system

3.1.5.2 Electric Circuit Design

Figure 3.7 shows the circuit design of the system which using the Arduino UNO board as a medium to connect the EMG signal conditioning, emergency switch and servo motor. EMG signal conditioning is connected at pin 0 and emergency switch is connected at pin 1. Both of the pin is used for analog and digital input.

Hip servo motor board is connect at pin 9 and knee servo motor board is connected at pin 10. Both of the pin is used for PWM connection. The servo motor board is already connected the servo motor when buy. The voltage source for the servo motor board is 24V and the Arduino board is 5v.

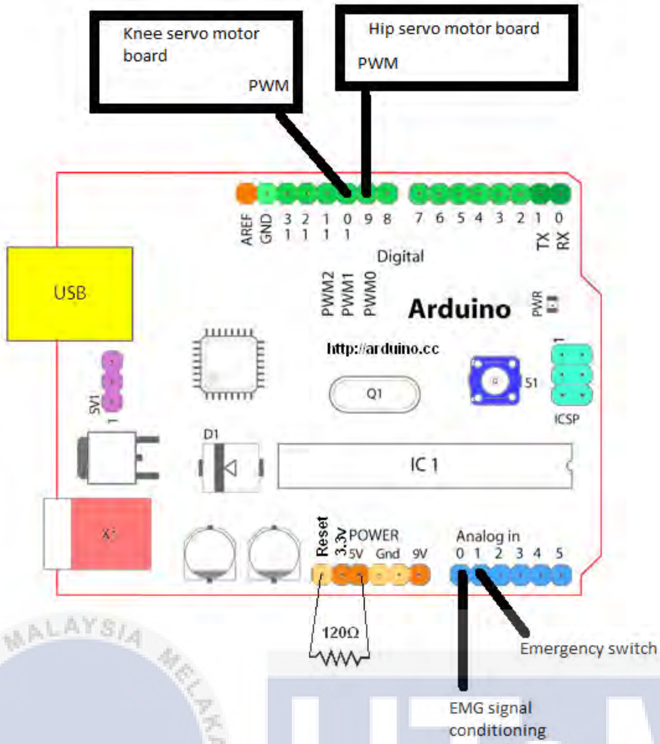

Figure 3.7: Circuit design of the system

3.2 EMG Sensor

3.2.1 Block Diagram of the EMG Sensor Circuit

Figure 3.8 shows the block diagram of the EMG sensor circuit. The sensor is used to detect the muscle signal. The signal conditioning is used to condition the signal so that the output signal is desired. The AgCl electrodes is used as the electrode to detect the signal from the muscle. In the signal conditioning stage, there are preamplifier circuit, filter circuit, noise cancellation circuit, inverting amplifier circuit, rectifier and smoothing circuit. After that, the output signal from the signal conditioning stage is displayed on the oscilloscope to ensure the output signal is desired.

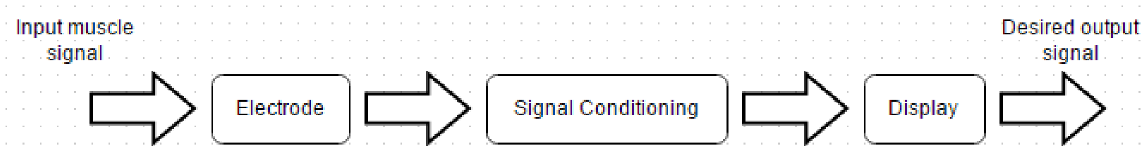

Figure 3.8: Block diagram of the EMG signal conditioning circuit

3.2.2 EMG Sensor Schematic Circuit

3.2.2.1 Type of Circuit of EMG Sensor

The schematic circuit is made using the Proteus software. The circuit contains of preamplifier circuit, driven right leg circuit, second order band pass filter, inverting amplifier, rectifier and a smoother. The AgCl electrode is used to detect the muscle signal from the lower limb.

The preamplifier circuit as shown in Appendices E is used to differentiate the input EMG signal. The capacitor is used to ensure the supply voltage will not drop. This Driven Right Leg as shown in Appendices E is used to reduce the noise produce by the body.

Frequency of the lower limb is 10Hz to 500Hz (Konrad, P [9]). To get the 10Hz and 500Hz range of frequency, a band pass filter as show in Figure 3.9 is used. The band pass filter consists of high pass filter and low pass filter.

For the high pass filter, the capacitor value is fixed at 100nF.

$$R = 1/(2\pi\ x\ f\ x\ C) \tag{3.9}$$

For low pass filter, the capacitor value is fixed at 10n.

$$R = 1/(2\pi\ x\ f\ x\ C) \tag{3.10}$$

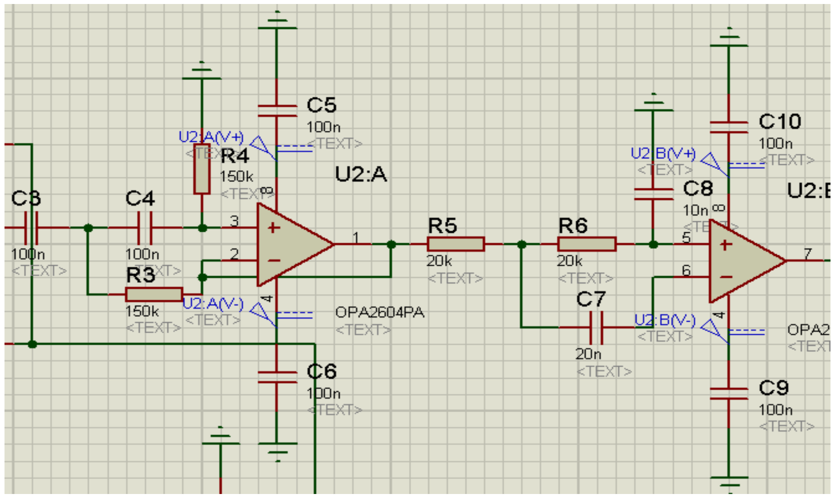

Figure 3.9: Band pass filter

The inverting amplifier as shown in Figure 3.10 is used to invert input signal.

$$Gain = -\left|\frac{R10}{R11}\right|, \text{ where gain} = 1 \tag{3.11}$$

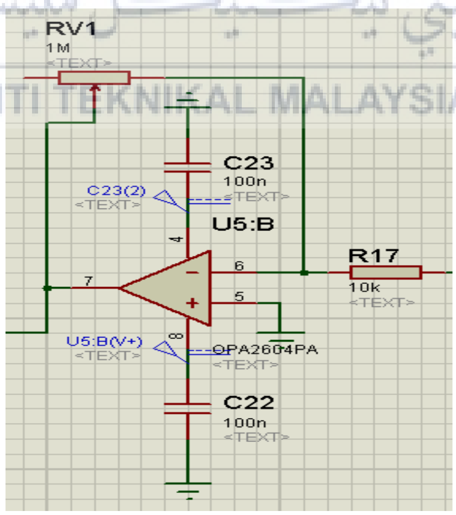

Figure 3.10: Inverting amplifier

The rectifier used in the circuit is high precision rectifier as shown in Figure 3.11 is to rectify the circuit.

According to Ken Bigelow [25], the resistor value can be calculated using the equation,

$$R13 = R19 = R18 \tag{3.12}$$

$$R14 = R12 = 2R13 \tag{3.13}$$

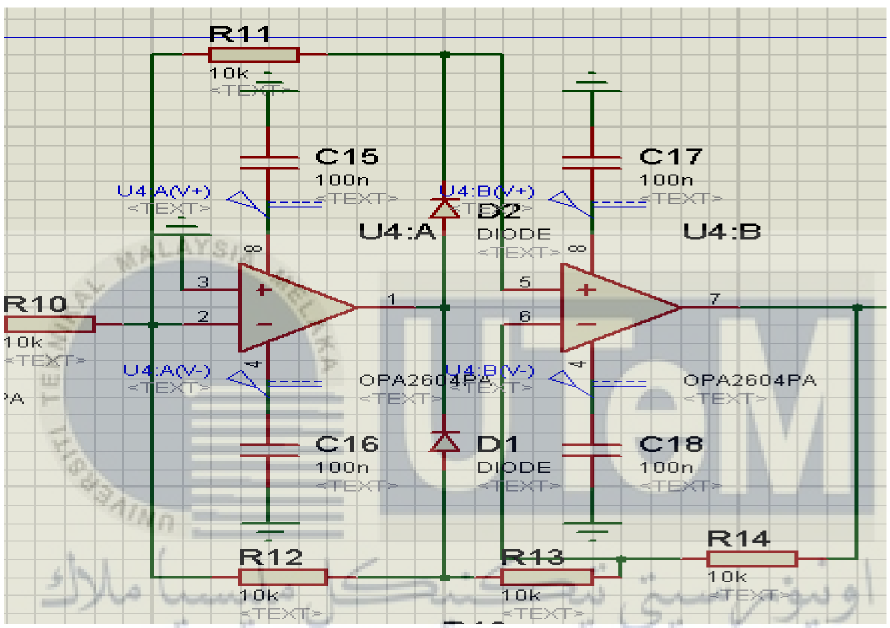

Figure 3.11: Rectifier

The smoother as shown in Appendices E is used to smooth the input signal so that the signal can become dc voltage. By combining the smoother and rectifier, the raw EMG signal become linear envelop. The purpose of these 2 circuit is to make the EMG signal become linear envelop so that feature extraction from the EMG signal can be done.

3.2.2.2 EMG Sensor Complete Circuit

Appendices F shows the complete circuit of the EMG sensor. The inputs are connected to the AgCl electrode which used to receive the EMG signal. The output will be connected to the oscilloscope to see the trend of the signal. Table 3.3 shows the resistor and capacitor value. The resistor value for high pass filter is calculated using equation 3.9 by fixing the capacitor value at 100nF. The resistor value of low pass filter is calculated using equation 3.10 by fixing the capacitor value at 10nF while the resistor value for inverting amplifier is calculated using equation 3.11 by fixing one of the resistor value at 10kΩ. The resistor, R13 of rectifier is fixed at 20kΩ and using the equation 3.12 and 3.13 to calculate the rest resistor value.

Table 3.3: Resistor value and capacitor value for amplifier

Amplifier	Resistor value (Ω)	Capacitor value (F)
High pass filter	150k	100n
Low pass filter	32k	10n
Inverting amplifier	10k	-
Rectifier	R13=R18=R19=20k R12=R14=40k	-

3.2.2.3 Simulation of EMG Sensor

After the schematic circuit diagram is made, the circuit is then simulated using Proteus which is as shown in Figure 3.12. The red and purple line is the sinusoidal input. The black line is the reference which show the zero voltage value and the purple line is the output of the circuit which is a dc voltage which the input signal is rectified and smoothed.

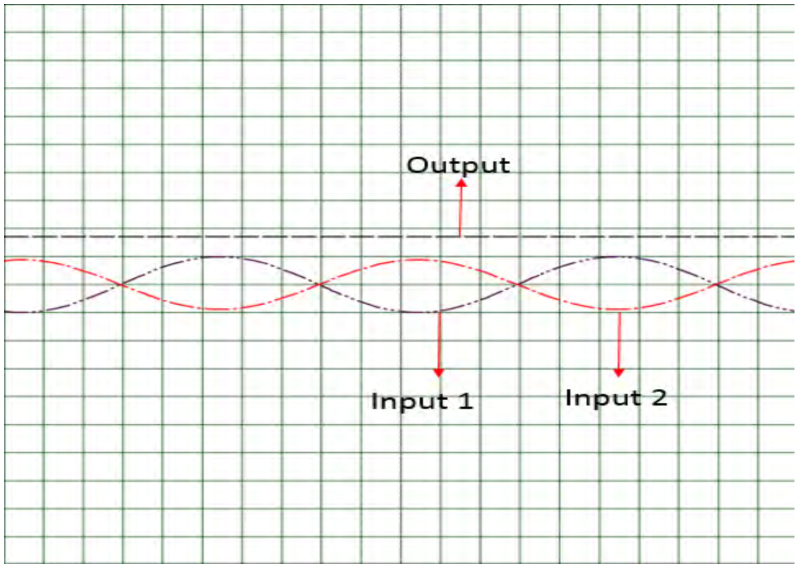

Figure 3.12: Simulation of the circuit

3.2.3 Constructed EMG Sensor Circuit

Figure 3.13 shows the constructed EMG sensor circuit which consists of preamplifier circuit, driven right leg, second order band pass filter, high precision rectifier, smoother and inverting amplifier. The constructed EMG circuit is tested before the experiment is carried by using the function generator and digital oscilloscope as shown in Figure 3.14.

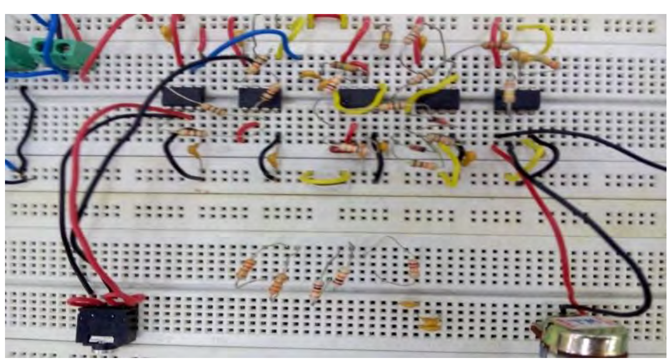

Figure 3.13: Constructed EMG sensor circuit

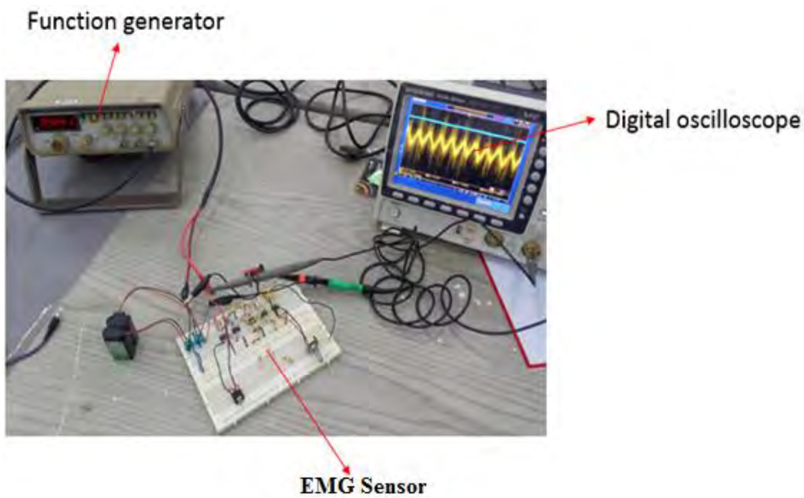

Figure 3.14: Testing the EMG sensor circuit using function generator

Figure 3.15 shows the graph from the EMG sensor circuit and function generator. The yellow line is the input from function generator to EMG sensor circuit which is a sinusoidal wave with noise. The green line is the output from EMG sensor circuit which is amplified, rectified and smoothed.

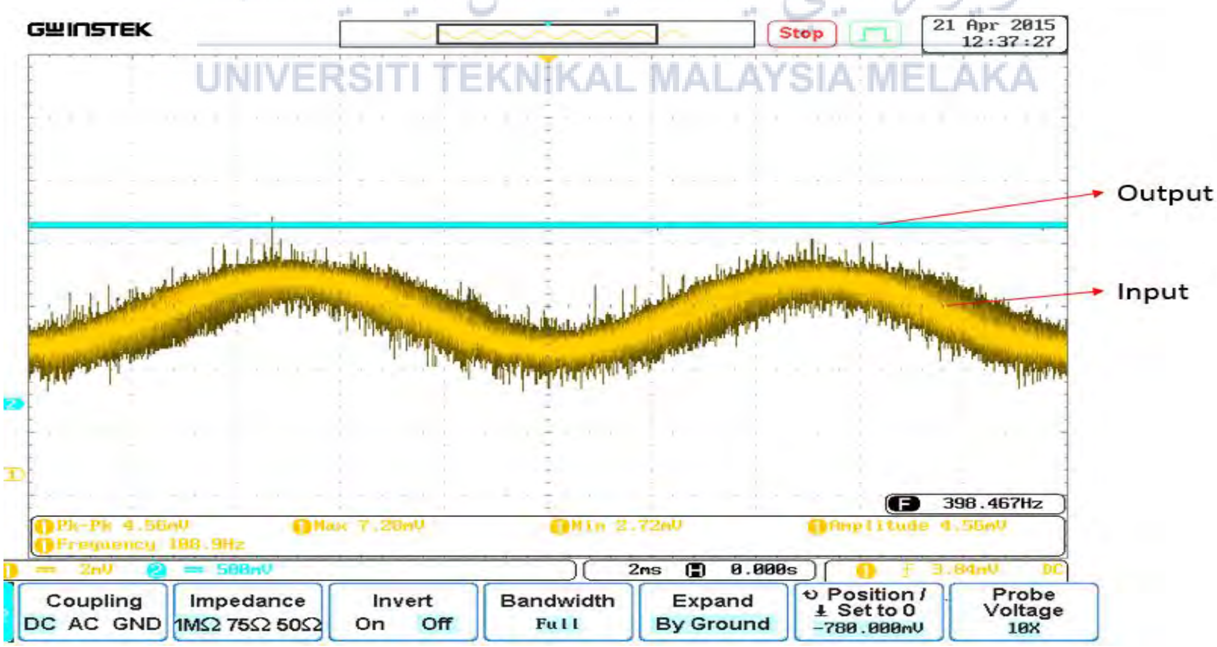

Figure 3.15: Signal Graph

3.3 Experiment Set Up for Performance Analysis

3.3.1 Experiment Set Up for Getting the EMG Signal Value Based on the Walking Motion

According to Sharif Muhammad Taslim Reza [23], the EMG signal from lower limb muscle is affected by the motion of the lower limb which means that the EMG signal is perpendicular to the motion of the lower limb. In his experiment, he recorded 5 subjects who are male with around 170 cm of height and 56 kg of weight. The subjects perform a sit and stand motion. Therefore, based on that article, a similar procedure is made.

This experiment objective is to study the behavior of EMG signal value based on the walking motion on 5 male subject which is around 166 cm of height and 56 kg of weight and find their average EMG value so that the value can be used to on and off the lower limb rehabilitation device.

Material:

1. EMG sensor
2. AgCl electrode
3. Cables
4. Digital oscilloscope

The procedures are:

1. The skin is cleaned at the rectus femoris.
2. The electrodes is placed at the muscle as shown in Figure 3.16.
3. The impedance of the muscle is measured and check based on the Table 3.4 whether it need to run a second clean or not.
4. After succeed, a simple warm up like walking is needed for at least 3 minutes.
5. Cables is connected with the electrodes.
6. The leg is Initial Contact motion as shown in Figure 3.17 and the EMG signal is recorded.
7. Walking motion for Loading Response phase is performed as shown in Figure 3.18.

8. The readings is recorded.
9. Step 1 until step 8 are repeated for 5 subjects.
10. The EMG signal value is analyzed by taking the maximum and minimum voltage value and find their average value and compared with the walking motion.

Precaution: Make sure the power cable of the digital oscilloscope is not share by other device so that there is no disturbance which could affect the accuracy of the result.

Table 3.4: Impedance range for EMG signal [9]

Impedance range (kΩ)	Recommendation
1 -5	Very good condition
5 - 10	Good and recommended if feasible
10 - 30	Acceptable for easy conditions
30 - 50	Less good, attention is needed
>50	Should be avoided or requires a second cleaning run

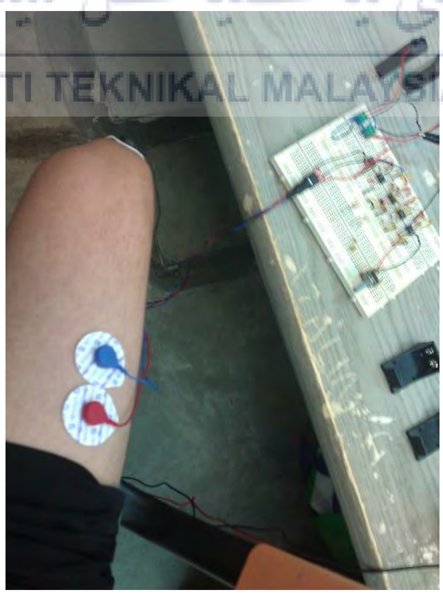

Figure 3.16: AgCl electrode is placed at rectus femoris muscle

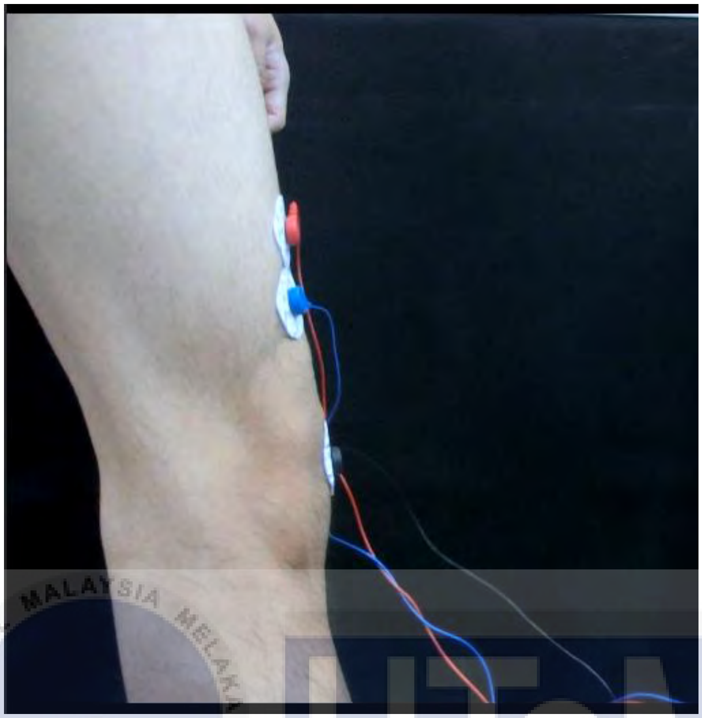

Figure 3.17: The leg is at Initial Contact phase

Figure 3.18: Walking motion at Loading Response phase

3.3.2 Experiment Set Up for Analyzing the Repeatability Performance of the Trajectory Planning of Lower Limb Rehabilitation System

According to Zhiqiang Zhang [26], the desired angle motion of the lower limb rehabilitation device can be calculated and compare it with the measured angle from the device. The measured value will have similar result with the desired angle. In his experiment, the walking motion of the device is repeated for 10 times. Therefore, based on that article, a similar procedure is made.

This experiment objective is to analyze the repeatability performance of the lower limb rehabilitation system by analyzing the trajectory of the device. The angles of the device of each joint are collected and compared with the desired value.

Material:

1. Sonxun SXSV50 servo motor with connected to the hardware of hip and knee joint
2. Arduino UNO circuit board
3. Wires
4. A piece of A4 paper
5. Protector
6. Toothpick
7. Lithium Ion battery 12V

The procedures are:

1. The reference point is drawn and set as shown in Figure 3.19 to increase the accuracy of the measurement so that every time the hardware is placed at the same place.
2. The circuit is constructed as shown in Figure 3.20.
3. The toothpick is placed at the hardware of hip joint as shown in Figure 3.21 so that it can be act as a point to measure the angle of the joint when the hardware rotate.

4. The hardware of the circuit is set up by placing the hardware on top of the reference point as shown Figure 3.21.
5. The hip joint hardware is placed at the center of reference point and the circuit is on.
6. The measurement is recorded for one complete gait cycle of the hip.
7. Step 4 until step 6 are repeated for 10 times.
8. Step 3 until step 7 are repeated by replacing the hip joint hardware with knee joint hardware on the reference point.

Precaution: Make sure the hardware of the joints in place at the exact center of the reference point by marking the point on the reference paper so that the joint place at the same time when carry out the experiment and the tooth pick is place properly by using the tape to stick to the hardware so that the tooth pick will not move every time the hardware rotate which could affect the accuracy of the result.

Figure 3.19: Reference point

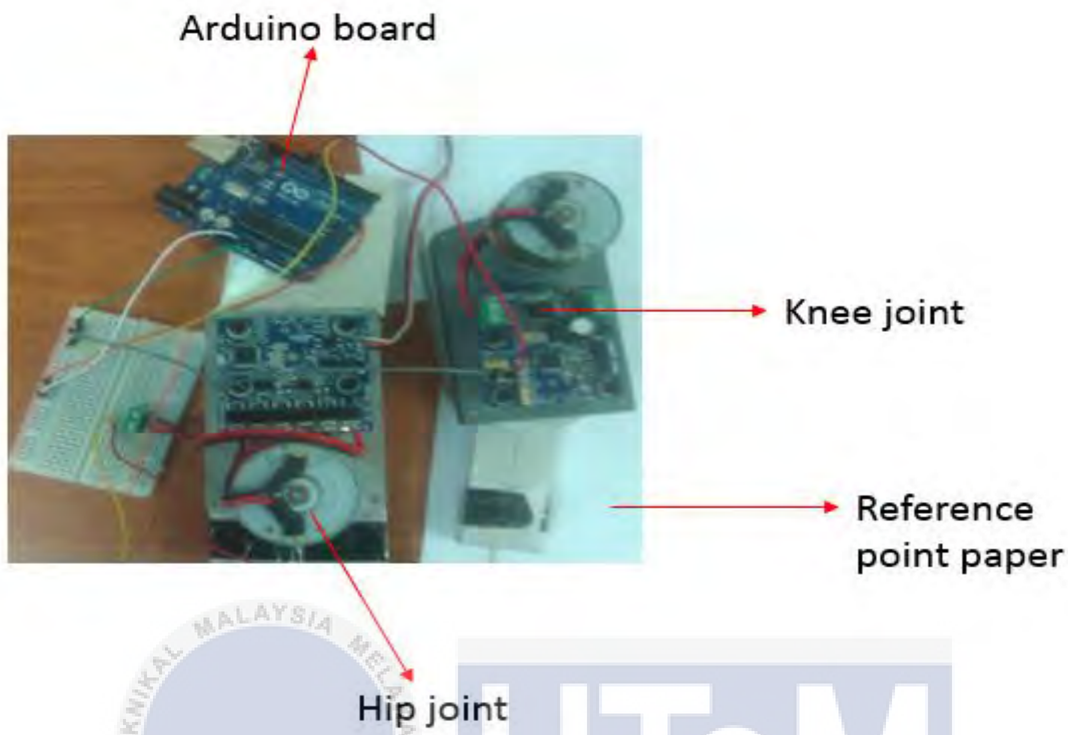

Figure 3.20: Circuit for the hardware

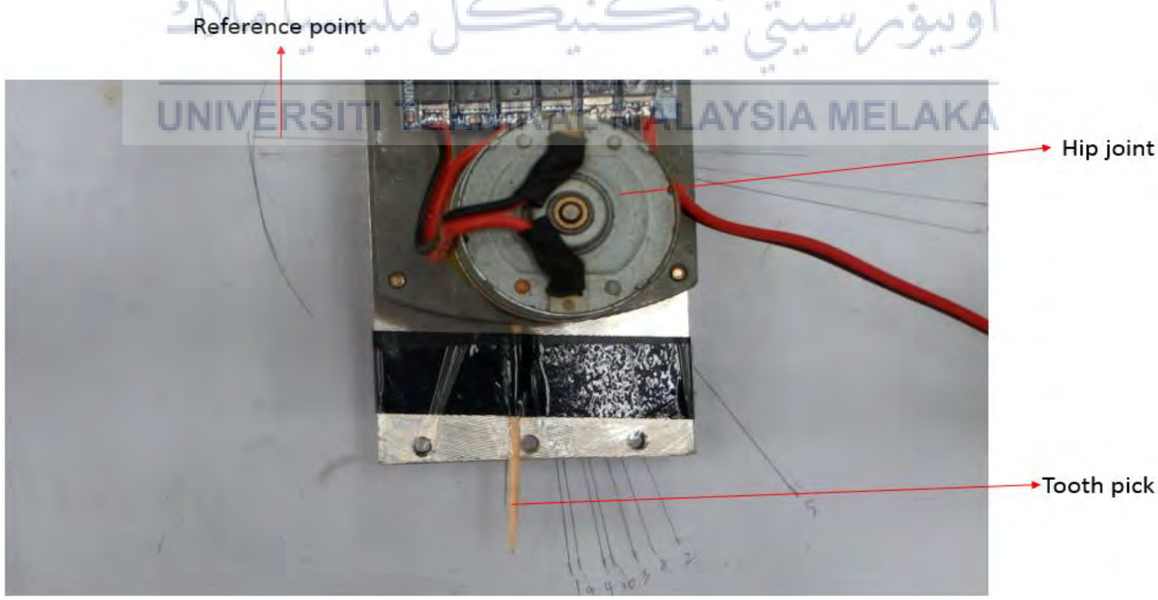

Figure 3.21: Setting of the hardware of hip joint

3.3.3 Experiment Set Up for Analyzing the Accuracy Performance of the EMG Sensor

According to Konrad, P [9], an accuracy analysis can be carried out by comparing the EMG signal from the own signal conditioning with desired EMG signal. In his experiment, a subject perform a sit and stand motion for 3 cycles. Therefore, based on that article, a similar procedure is made. The muscle chosen is rectus femoris. This experiment objective is to analyze the accuracy performance of the EMG sensor and compare the value with the muscle sensor.

Material:

1. Muscle sensor V3
2. EMG sensor
3. AgCl electrode
4. Cables
5. Digital oscilloscope

The procedures are:

1. The cleaning and warm up process is carried out exactly same as the Experiment 3.3.1 step 1 until step 8.
2. Step 1 is repeated for 3 times.
3. The muscle sensor V3 is used and step 2 and step 3 are repeated.
4. Both readings is analyzed by comparing their peak to peak voltage value of each of the circuit.

Precaution: Make sure the power cable of the digital oscilloscope is not share by other device so that there is no disturbance which could affect the accuracy of the result and make sure the electrode is placed at the same place of the muscle when using EMG sensor and Muscle sensor.

Chapter 4

RESULT AND DISCUSSION

This chapter covers all the experimental result and discussion of the experiment. This section also covers the conclusion of the experiment.

4.1 EMG Signal Value Data Collecting Based on the Walking Motion

The objective of this experiment is to study the behavior of EMG signal value based on the walking motion on 5 male subject. Table 4.1 shows the maximum and minimum voltage of experiment result getting from 5 male subject by using the EMG sensor.

The experimental subjects who are male around 56 kg and have a height of 166cm are carried out the experiment for collecting his walking motion data. The experiment is carried out with 5 subjects as shown in Table 4.1. Table 4.2 shows the average maximum voltage of the 5 subjects is 0.512V and average minimum voltage is -0.2856V.

The data of EMG signal is collecting based on the first and second walking gain phases due to the complexity of collecting the whole walking gain phases data. The average behavior of the subjects with the same characteristic EMG value based on the first and second walking motion is 0.512V for maximum and -0.2856V for minimum. The average of the maximum and minimum value can be implemented in the arduino program code which as shown in Appendices D.

Therefore, the objective of this experiment is achieved. The accuracy of the average general EMG value can be increased by increasing the number of subject with same characteristic.

Table 4.1: Experiment result of EMG sensor when perform walking motion

Subject	Maximum voltage (V)	Minimum voltage (V)
1	0.5	-0.26
2	0.56	-0.28
3	0.44	-0.348
4	0.56	-0.28
5	0.5	-0.26

Table 4.2: Average voltage value for experimental subject

Average maximum (V)	Average minimum (V)
0.512	-0.2856

4.2 Repeatability Performance of the Trajectory Planning of Lower Limb Rehabilitation System

This experiment objective is to analyze the repeatability performance of the lower limb rehabilitation system by analyzing the trajectory of the device. According to Murray MP [27], the average time for a healthy human walking for one step is 1.03 seconds. By using the equation 3.6, 3.7 and 3.8 to find the value of a0, a2 and a3 for finding the general equation of the line. Table 4.3 shows the value of a0, a2 and a3 for hip and knee joint at each phase.

After finding the value of a0, a2 and a3, by using the equation of 3.4 to plot the graph of the motion of hip and knee joint for one gait cycle which is as shown in Figure 4.1 and Figure 4.2. The results show the trend of motion of hip and knee joint which are same as the trend as shown in Figure 3.4. However, the graph in Figure 3.4 is using the degree of motion versus percentage of gait cycle but Figure 4.1 and 4.2 are using the degree of motion versus time. The calculated of the desired trajectory planning is shown in the Appendices G.

Appendices H shows the measurement of the hip joint angle and Appendices I shows the measurement of the knee joint angle which both are in radian. Both Appendices H and I contain the desired angle and average angle which calculated from the measurement of hip and knee joint angle.

The unit from result of mean square error is squared compare to root mean square error. This cause the result of root mean square error is more easily to understand. Therefore by using the root mean square error formula, average angle formula and excel function to calculate the confidence level which is used to find the consistency of the hardware:

$$avaerage = (\sum_{i=1}^{n} a_n)\,/10 \tag{4.1}$$

$$RMS\ error = \sqrt{\frac{(\sum_{i=1}^{n}(a_n - desired\ value)^2)}{number\ of\ sample}} \tag{4.2}$$

Table 4.3: The value of a0, a2 and a3 for each joint at each phase

	Gait cycle (%)										
	IC (0%)	**LR (0%-10%)**	**MST (10%-30%)**	**TST (30%-50%)**		**PSW (50%-60%)**	**ISW (60%-73%)**		**MSW (73%-87%)**	**TSW (87%-100%)**	
	0%	**10%**	**30%**	**40%**	**50%**	**60%**	**66.50%**	**73%**	**87%**	**93.50%**	**100%**
time (sec)	0	0.13	0.309	0.412	0.515	0.618	0.68495	0.7519	0.8961	0.96305	1.03
time interval (sec)	0	0.13	0.179	0.103	0.103	0.103	0.06695	0.06695	0.1442	0.06695	0.06695
a0 (hip)	30		30			-10				35	
a2 (hip)	0		-809.58			929.5139				-836.6236	
a3 (hip)	0		1401.8702			-1626.0192				4165.415	
a0 (knee)	5		15		5			60			0
a2 (knee)	1775.1479		-377.2446		2214.7145			-2327.2973			3346.4943
a3 (knee)	-9103.3227		891.8312		-5409.3289			5579.2816			-33323.3194

Based on Table 4.4 and Table 4.5, the root mean square error for all phases are smaller than 1 which is relatively small. The highest value of root mean square error for the hip joint is 0.2699 radian which is at Terminal Stance 2 phase while the highest value of root mean square error for the knee joint is 0.4037 radian which is at Initial Swing 2 phase. The confidence level for hip joint of each phase is between 0.0156 radian and 0.0612 radian while the confidence level for knee joint of each phase is between 0.0161 radian and 0.0377 radian. This shows that the hip joint and knee joint have very high consistency and accuracy.

Table 4.4: Hip joint RMS error and confidence level

Hip angle (degree)		
Phases	**RMS error**	**Confidence level (95%)**
Initial Contact	0.1118	0.0612
Loading Response	0.1118	0.0612
Mid Stance	0.0761	0.0410
Terminal Stance 1	0.1942	0.0156
Terminal Stance 2	0.2699	0.0179
Pre Swing	0.2225	0.0169
Initial Swing 1	0.1786	0.0179
Initial Swing 2	0.1668	0.0179
Mid Swing	0.0691	0.0222
Terminal Swing 1	0.0603	0.0234
Terminal Swing 2	0.0774	0.0175

Table 4.5: Knee joint RMS error and confidence level

Knee angle (degree) / Phases	RMS error	Confidence level (95%)
Initial Contact	0.0529	0.0377
Loading Response	0.0518	0.0262
Mid Stance	0.0515	0.0222
Terminal Stance 1	0.0317	0.0208
Terminal Stance 2	0.1525	0.0202
Pre Swing	0.3935	0.0161
Initial Swing 1	0.3050	0.0161
Initial Swing 2	0.4037	0.0169
Mid Swing	0.1235	0.0244
Terminal Swing 1	0.0830	0.0306
Terminal Swing 2	0.0611	0.0289

Figure 4.1 and Figure 4.2 shows the pattern of the graph of desired angle is same as the measured angle. However, there is some difference between the desired angle and measured angle and deviate more when the angle is at its peak value as shown in the Figure 4.1 Box A and Figure 4.2 Box B. This is due to the shaft of the servo motor does not properly fit in the joint and cause the accuracy of the servo motor reduce. This means that the hole that used to place the shaft of the servo motor is too big and cause the shaft of the servo motor slips every time it rotate.

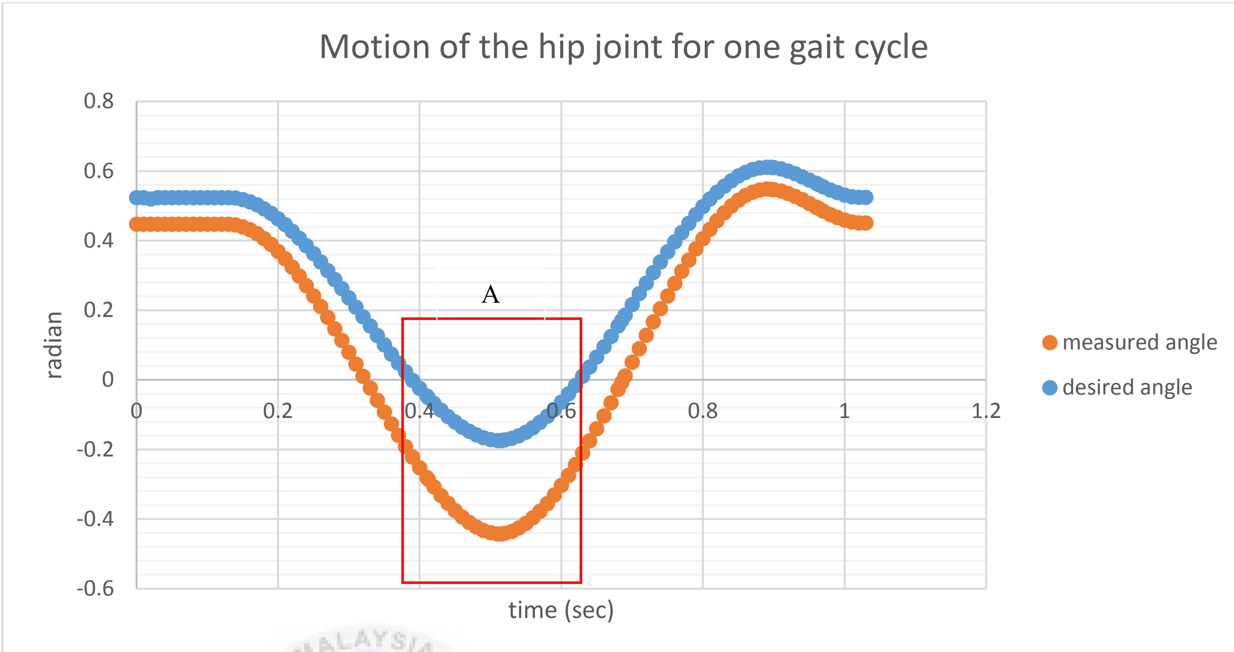

Figure 4.1: Graph of the desired and measured hip joint angle

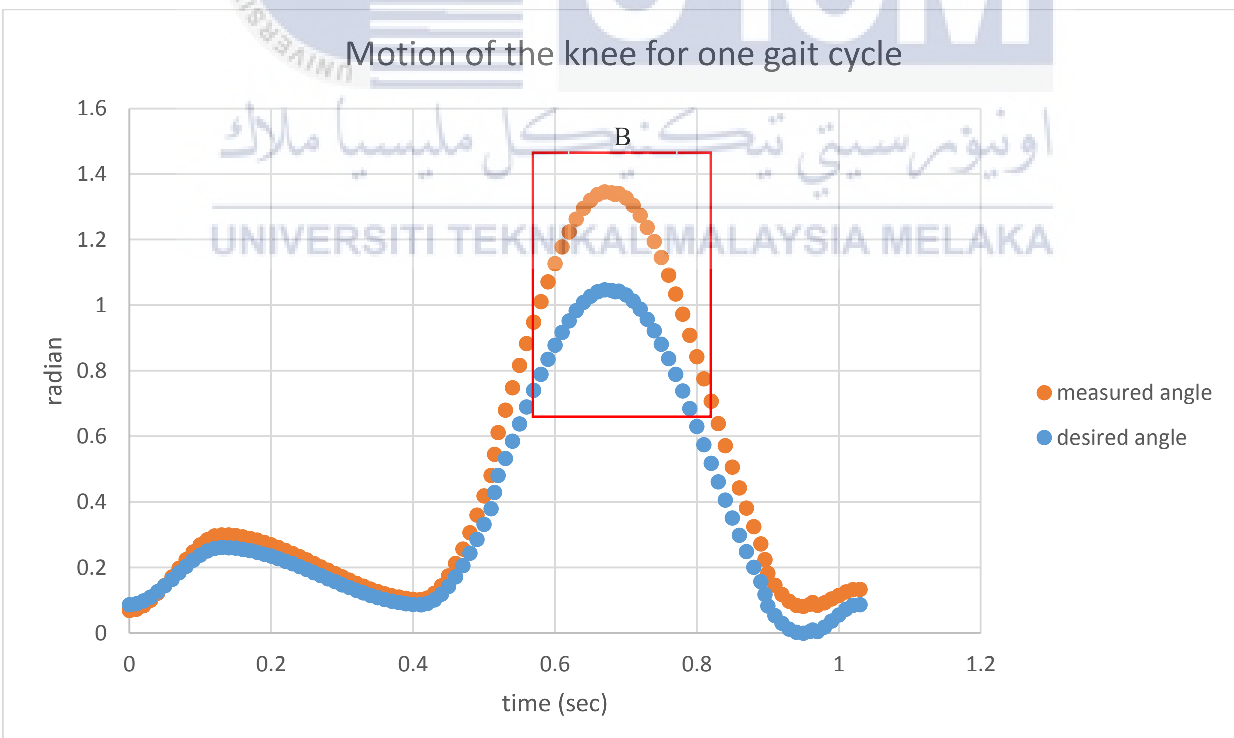

Figure 4.2: Graph of the desired and measured knee joint angle

The lower limb rehabilitation device having a relatively small error and relatively high consistency. This shows that the performance of the lower limb rehabilitation device is high. Therefore, it can be used on the post stroke patients in order to help them to regain their walking ability. As a conclusion, the objective of this experiment is achieved.

In order to increase the accuracy and reduce the error of the device, some improvement of the device need to be taken. The accuracy of the device can be improve by replacing the hardware of joint that properly fit the shaft of the servo motor.

4.3 Analyze the Performance of the EMG Sensor

This experiment objective is to analyze the accuracy performance of the EMG sensor and compare the value with the muscle sensor. Table 4.6 shows the experiment result of muscle sensor and EMG sensor by recording their peak to peak voltage.

Table 4.6: Experiment result of muscle sensor and. EMG sensor

	Peak to peak voltage (V)		
Circuit	1	2	3
EMG Sensor	0.76	0.76	0.78
Muscle Sensor V3	0.74	0.73	0.76

By using the accuracy equation,

$$accuracy = 100 - \frac{Theoretical\,Value - Measured\,Value}{Theoretical\,Value}\,x\,100\% \qquad (4.3)$$

Based on the results show in Table 4.7, the average voltage of EMG sensor is 0.77V while the muscle sensor V3 is 0.74V. The muscle sensor V3 having a lower voltage than EMG sensor when perform same motion during experiment. The accuracy of the EMG sensor is around 96.86% when compare to the muscle sensor V3 as shown in Table 4.8.

Table 4.7: Average peak to peak voltage of muscle sensor and EMG sensor

Circuit	Average (V)
EMG Sensor	0.766666667
Muscle Sensor V3	0.743333333

Table 4.8: Accuracy of EMG sensor

Circuit	Accuracy (%)
EMG Sensor	96.86098655

The EMG sensor shows a very smooth graph as shown in Figure 4.3 Box C. The muscle sensor V3 shows there is a very high peak graph as shown in Figure 4.4 Box D. This shows that the EMG sensor shows a smoother graph than muscle sensor.

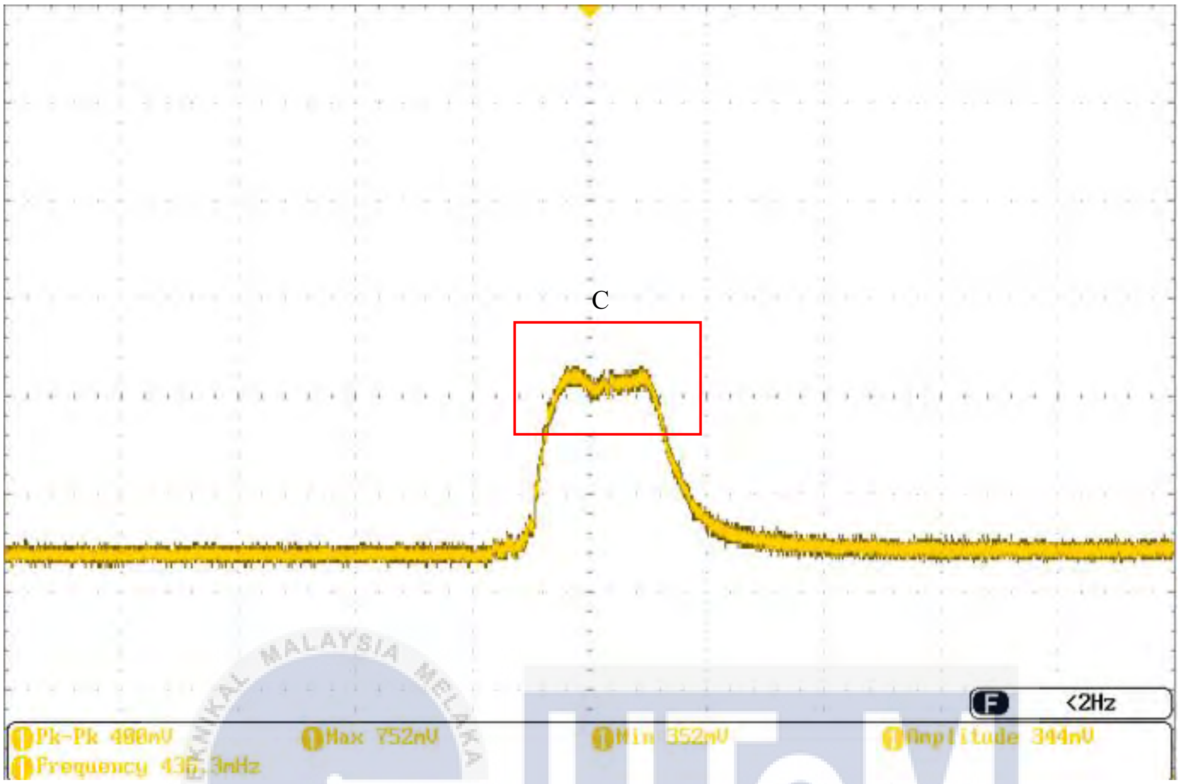

Figure 4.3: Example of experiment result from EMG sensor

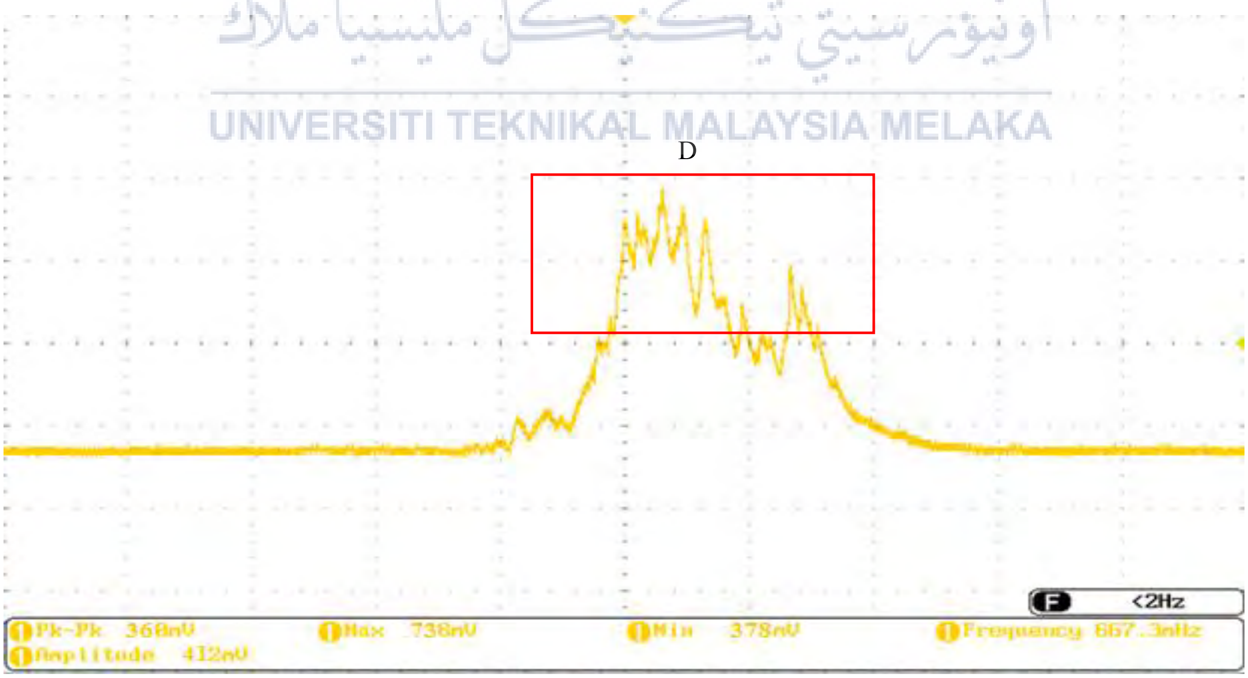

Figure 4.4: Example of experiment result from muscle sensor V3

Table 4.9 shows the characteristic of EMG sensor and muscle sensor. Both of the circuit is linear envelop type EMG circuit. However, the muscle sensor do not have bandpass filter and driven right leg. Muscle sensor is using the AD8221 as the IC at the preamplifier circuit which is purposely made for medical application and is high accuracy and very low noise disturbance while the EMG sensor using the AD620 as the IC at the preamplifier circuit which is cheap and is general purpose IC but high accuracy. This causes the EMG sensor having more noise disturbance compare to muscle sensor and this causes the EMG sensor having a higher value of peak to peak voltage compare to muscle sensor. Therefore, the accuracy of the EMG sensor is lower than the muscle sensor.

Table 4.9: Characteristic of EMG sensor and muscle sensor

Characteristic	**EMG Sensor**	**Muscle Sensor V3**
Type	Linear envelop	Linear Envelop
Preamplifier circuit	Yes	Yes
Bandpass filter	Yes	No
Driven right leg	Yes	No
Inverting amplfier	Yes	Yes
High precision rectifier	Yes	Yes
Smoother	Yes	Yes
IC used	AD620, OPA2604	AD8221, TL084

The EMG sensor is using the OPA2604 IC which is made for medical application and having a high accuracy, faster response and lower noise compare to TL084 which is general purpose IC used by the muscle sensor. This causes the EMG signal conditioning circuit having a smoother graph compare to the muscle sensor.

The accuracy of EMG sensor is relatively high when comparing to the muscle sensor V3. The reduction in the accuracy of the circuit is due to different component used in the circuit. However, the EMG sensor shows a smoother graph than muscle sensor. With these high accuracy and smoother graph, the EMG sensor can act as a Muscle sensor and the value getting from the EMG sensor can be used in the application. Therefore, the objective of this experiment is achieved. The accuracy of EMG signal conditioning can be improved by using a high precision and fast response amplifier IC.

Chapter 5

CONCLUSION AND FUTURE WORK

This chapter discuss about the conclusion based on the objectives of this project. This chapter also discuss about the recommendation that can be done in the future of the project.

5.1 Conclusion

The objectives of this project is to design a cubic polynomial and EMG based trajectory planning lower limb rehabilitation device for the post stroke patient to regain their walking ability and to analyze overall system performance in terms of accuracy and repeatability. In order to achieve these objectives, three experiments are conducted.

The first experiment is conducted to study the behavior of EMG signal value based on the walking motion on 5 male subject. From the study, we can see that the average behavior of the male subjects with the characteristic of around 166 cm of height and 56 kg of weight EMG value based on the first and second walking motion is 0.512V for maximum and -0.2856V for minimum. The average EMG value can be put inside the controller as the on off switch of the device.

The second experiment is conducted to analyze the repeatability performance of the lower limb rehabilitation device by analyzing the trajectory of the device. From the experiment, we can see that the performance of the lower limb rehabilitation device is high with a relative small root mean square error which is 0.2699 radian and 0.4037 radian and relative high consistency which is in the range of 0.0156 radian and 0.0612 radian. This shows that the performance of the lower limb rehabilitation device can be accepted and used in this project.

The third experiment is conducted to analyze the accuracy performance of the EMG sensor and compare the value with the muscle sensor. From the experiment, we can see that the performance of the EMG sensor is also high with the accuracy of the EMG sensor is around 96.86% when compare to the muscle sensor V3 relatively high accuracy compare with muscle sensor V3. This shows that the EMG sensor can be used in the lower limb rehabilitation system due to its high performance.

As a conclusion of this project, with the high performance of lower limb rehabilitation device and EMG sensor, the overall of the cubic polynomial and EMG based trajectory planning lower limb rehabilitation system performance is good and is suitable to be used on the post stroke patients in order to help them to regain their walking ability. Therefore, the lower limb rehabilitation system is successfully implemented and tested.

5.2 Future Work

This project can be further improved by including the ankle in the device. This means that the lower limb rehabilitation device will be a 3 DOF which cover hip, knee and ankle. This can help the post stroke patients who are unable to move their hip, knee and ankle. By having the actuator at the hip, knee and ankle joint, the device can help the post stroke patients to perform training at hip, knee and ankle joint. This project can help more post stroke patients in order to help them to regain their walking ability.

This project also can be improved by collecting the data for a full walking gait cycle. This can allow the device perform the walking gait by mapping the EMG signal with angle of the device.

This can allow the patients to stop at their desired angle or desired phase at any time. This can provide a more freedom to the patients rather than just follow the instruction of the device.

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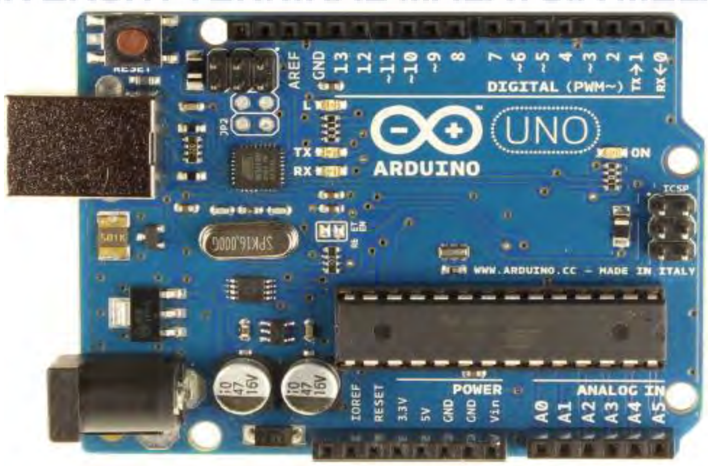
APPENDICES

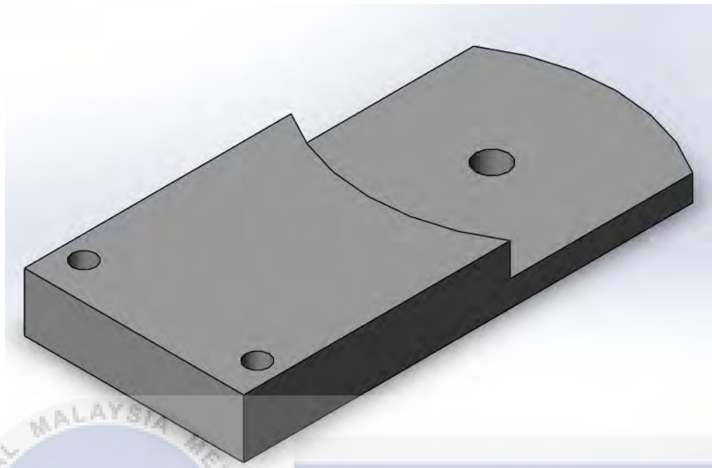
Appendices A: Component Used

Sonxun SXSV50 12 - 24 V High Power Servo Motor

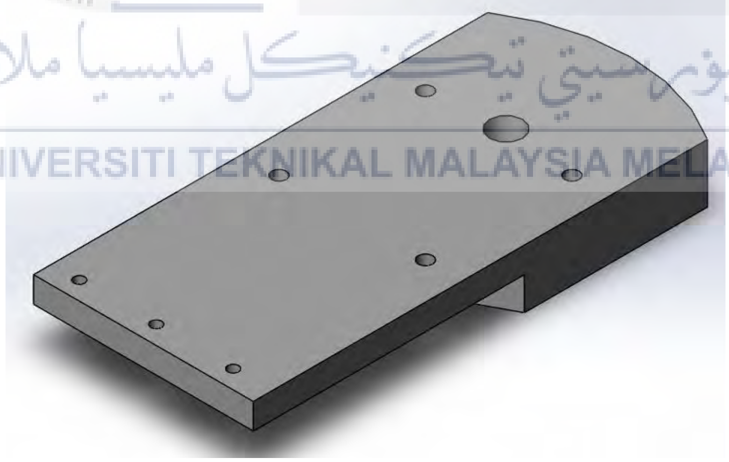


Arduino UNO board

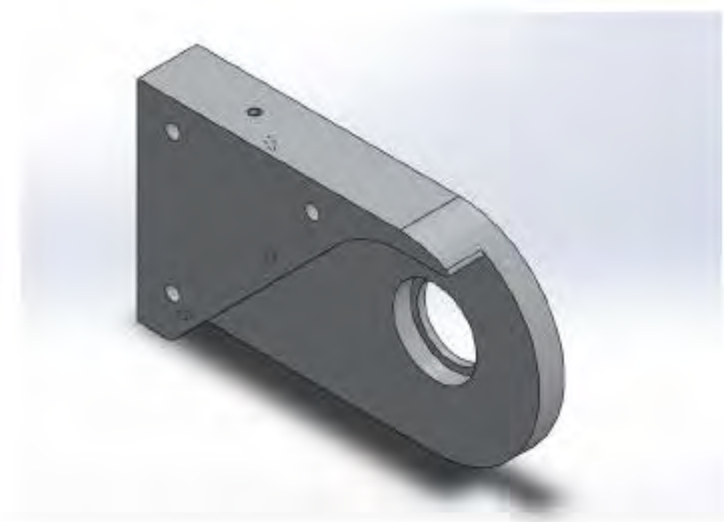
Appendices B: Lower Limb Design Drawing



Lower hip joint

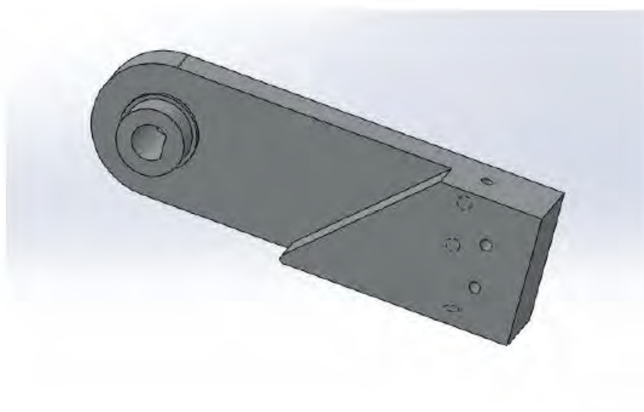


Upper hip joint



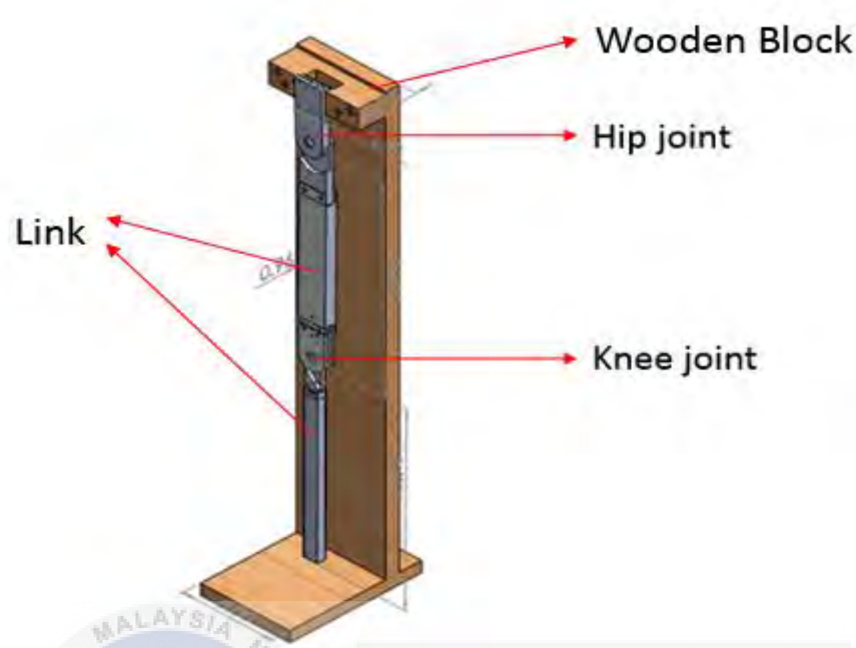
Upper knee joint

Motor plate



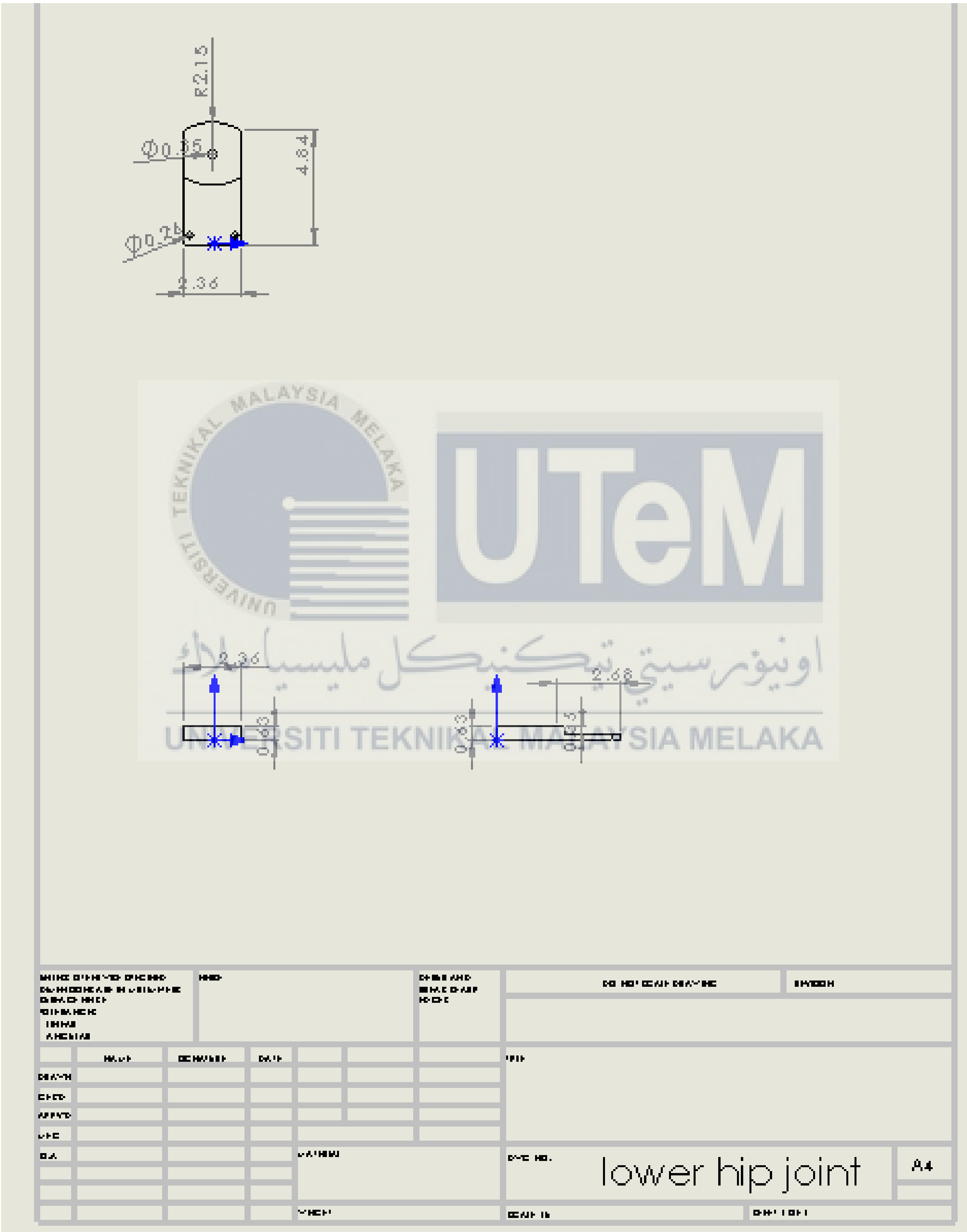
Lower knee joint

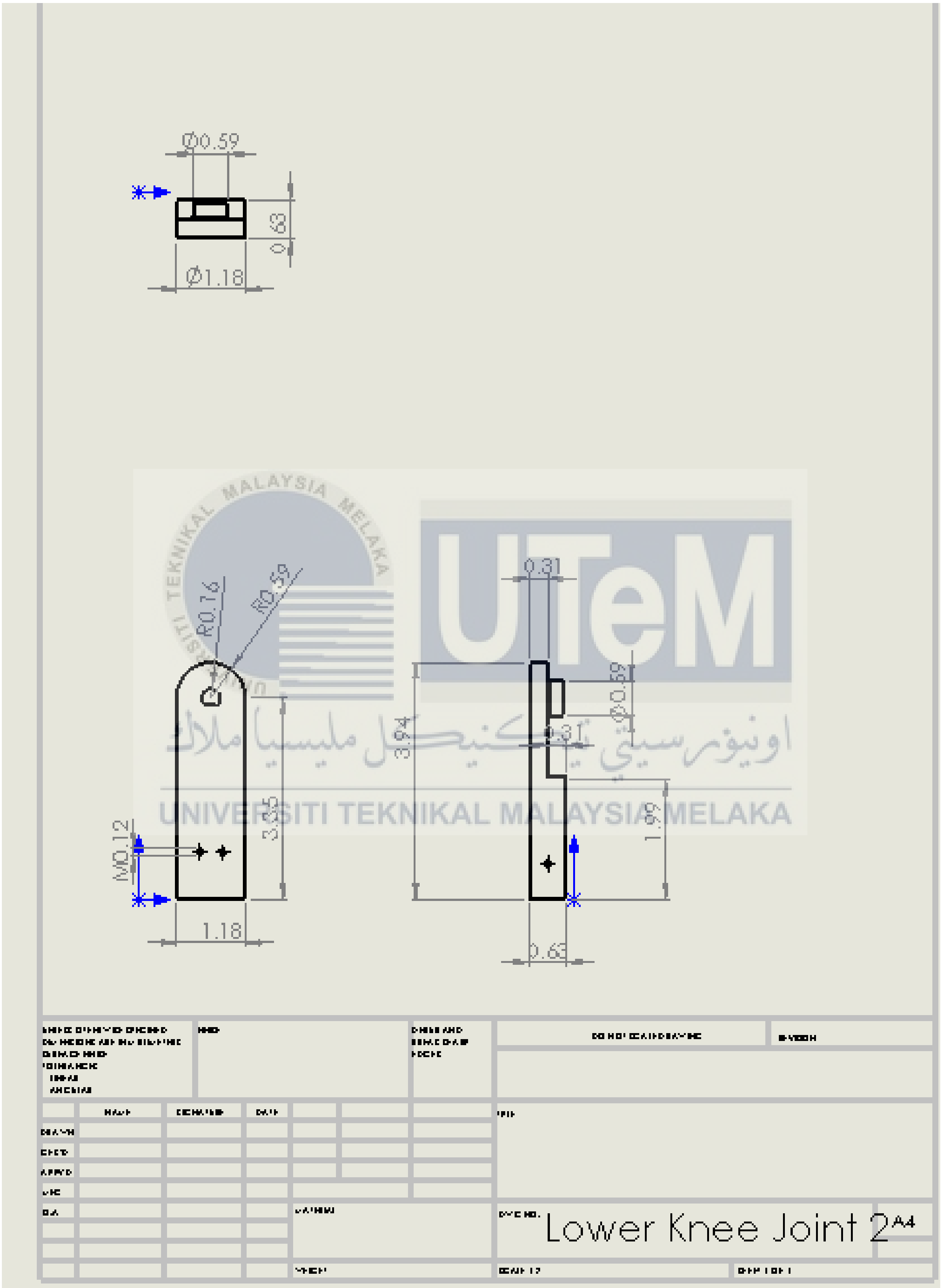
Knee joint connector

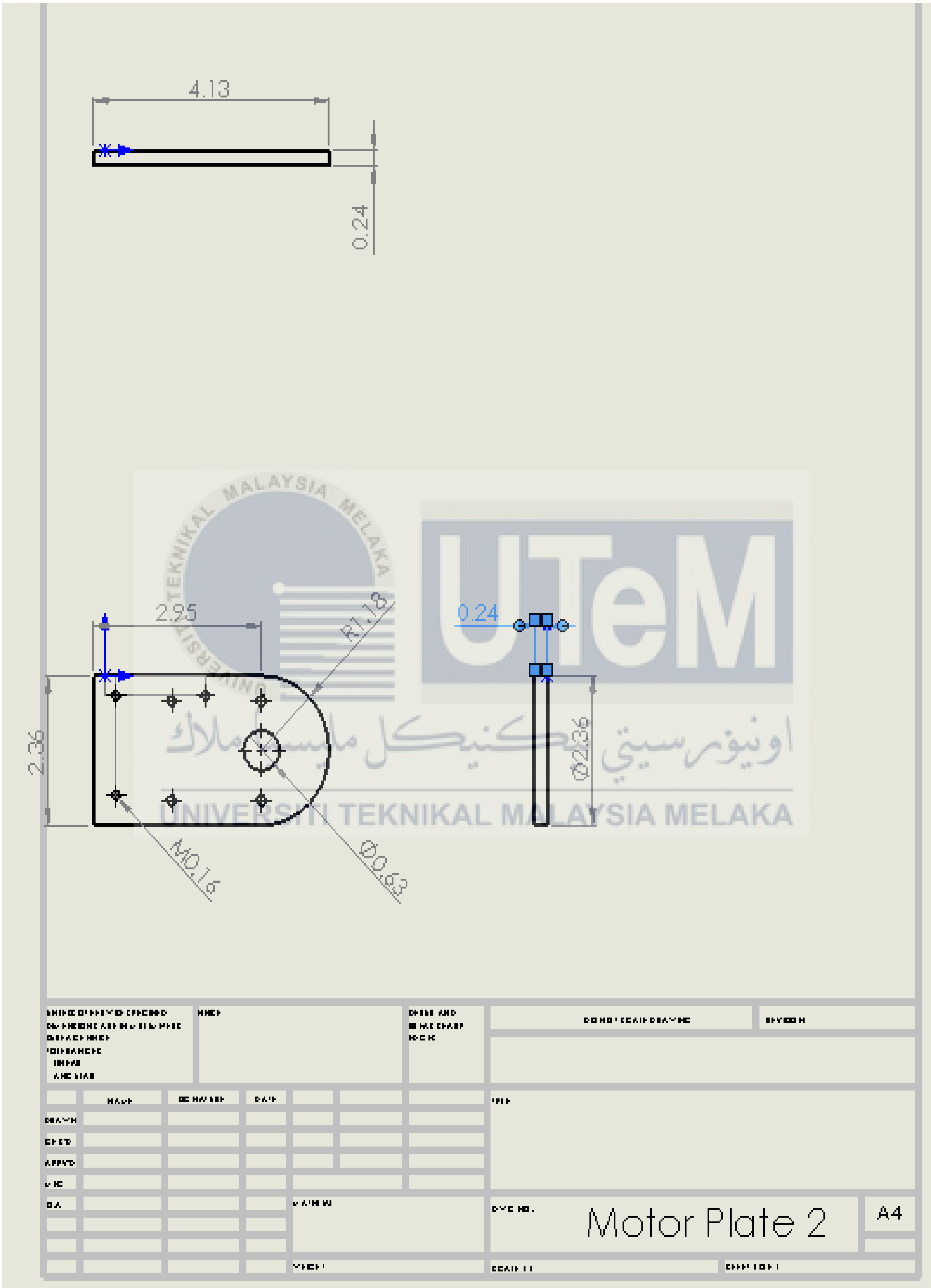


Assembly drawing of the device

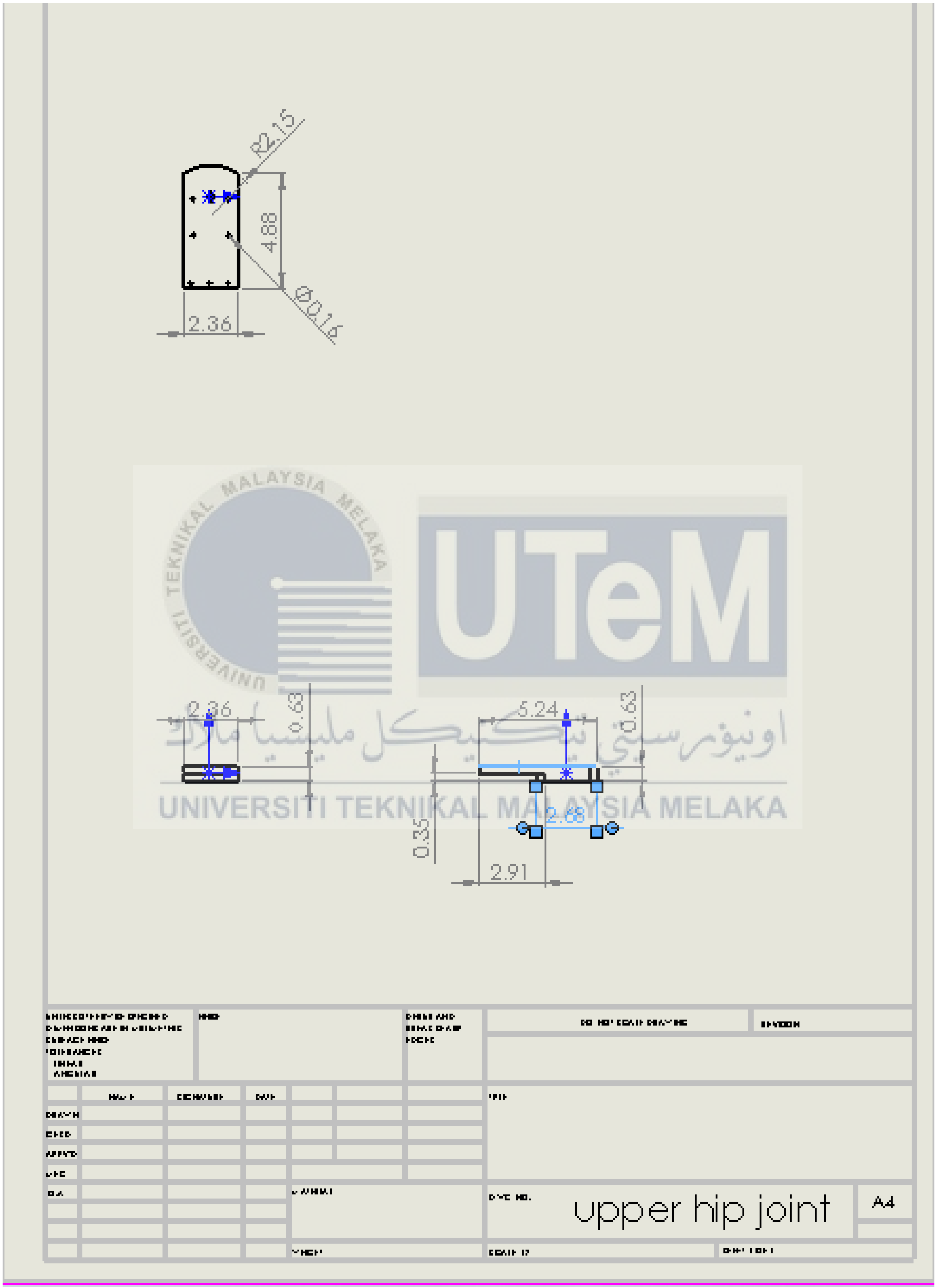
Appendices C: Dimension of Hardware Drawings

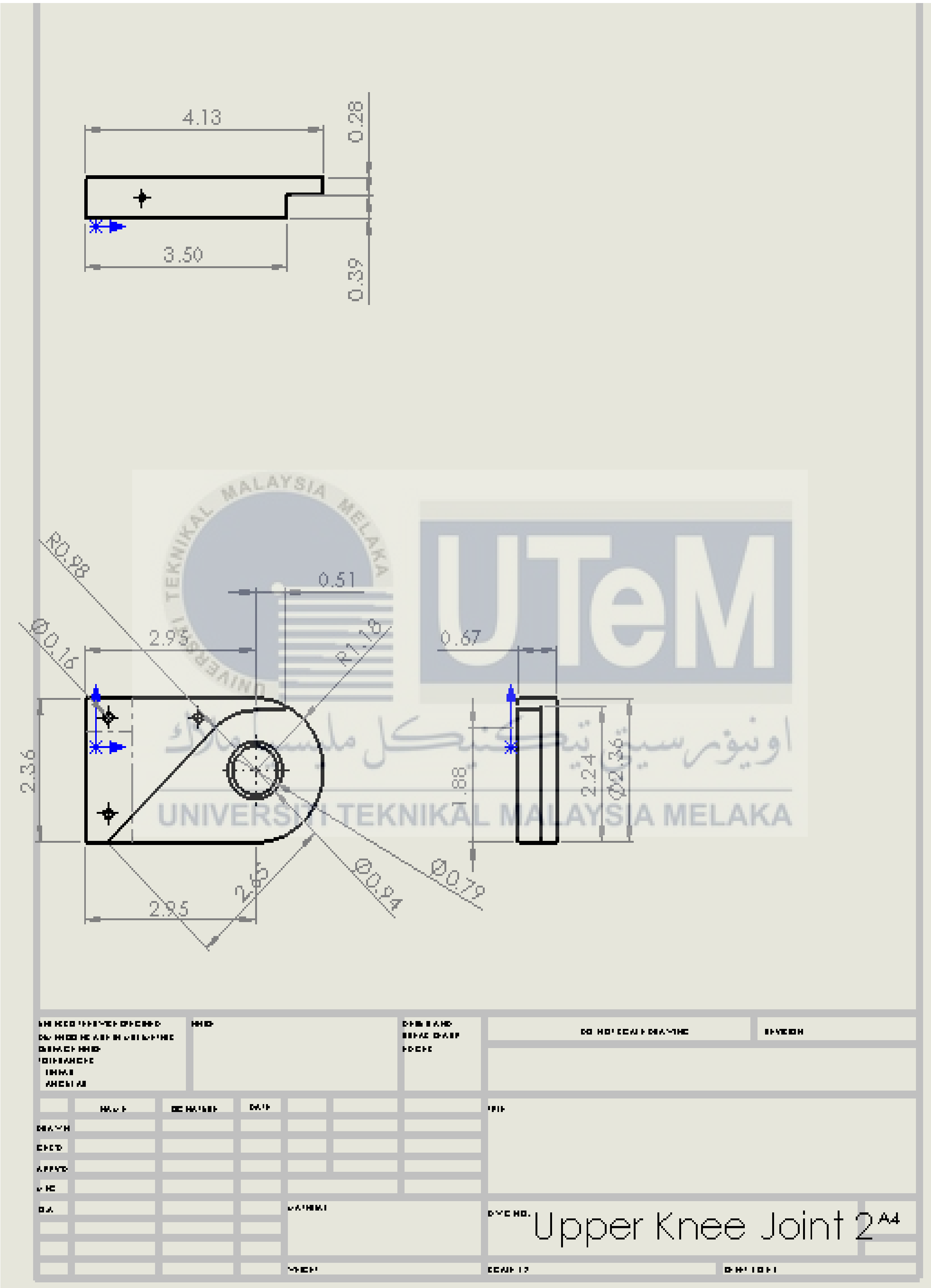



Ø0.59
0.63
Ø1.18
R0.16
R0.59
0.31
3.94
Ø0.59
0.31
3.35
1.99
M0.12
1.18
0.63
UNIVERSITI TEKNIKAL MALAYSIA MELAKA
UTeM
Lower Knee Joint 2
A4



4.13
0.24
2.95
R1.18
0.24
2.36
Ø2.36
M0.16
Ø0.63
UNIVERSITI TEKNIKAL MALAYSIA MELAKA
Motor Plate 2
A4


R2.15
4.88
Ø0.16
2.36
2.36
0.63
5.24
0.63
2.68
0.35
2.91
upper hip joint
A4


4.13
0.28
3.50
0.39
R0.98
Ø0.16
2.95
0.51
R1.18
0.67
2.36
1.88
2.24
Ø2.36
2.95
2.65
Ø0.94
Ø0.79
Upper Knee Joint 2
A4

Appendices D: Arduino Program Code

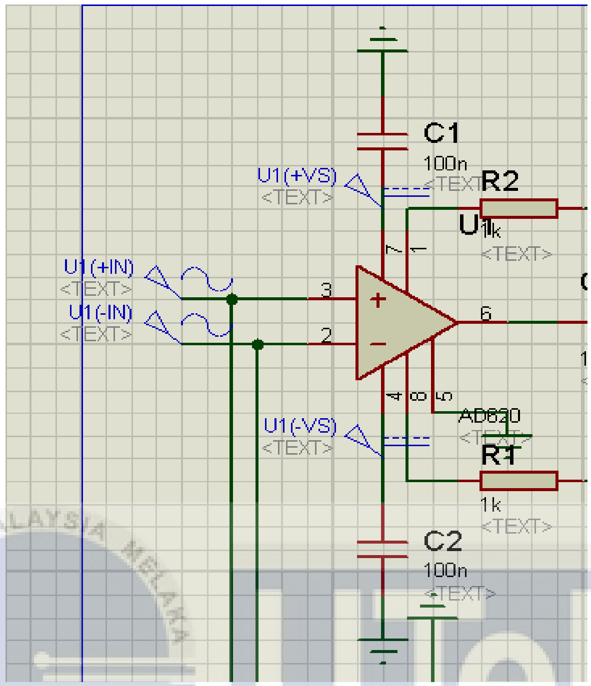
```
#include <Servo.h>
Servo hipservo;  //servo at hip joint
Servo kneeservo;  //servo at knee joint
int hipangle = 0;  //initial hip joint angle
int kneeangle = 0;  //initial knee joint angle
const int emg = 1;   //EMG signal at analog pin 0;
double sensor = 0;
double sensor 1;
int inputPin = 2; // input
double count = 0; //count time (ms)

void setup() {
  hipservo.attach(9);  //hip servo motor at pin 9
  kneeservo.attach(10);  //knee servo motor at pin 10
  pinMode (inputPin, INPUT);
  digitalWrite (inputPin, HIGH);
  Serial.begin(9600);
}

void loop() {
 sensor = analogRead(emg);
```

```
 Serial.println(sensor);
 hipservo.write(91);
 kneeservo.write(65);
sensor1 = sensor *9/1023
if (sensor1 <= 0.512 && sensor1 > 0)
  {
    hipservo.attach(9,500,2500);
    kneeservo.attach(10,500,2500);
    hipservo.write(127);  //IC
    kneeservo.write(78);  //IC
    delay (5000);
    hipangle = 30 + 97;
    kneeangle = 78;//**
    hipservo.write(hipangle);
    kneeservo.write(kneeangle);
    delay (1300);
    hipangle = 12 + 97;
    kneeangle = 78;//**
    hipservo.write(hipangle);
    kneeservo.write(kneeangle);
    delay (1790);
    hipangle = -3 + 97;
    kneeangle = 78;//**
```

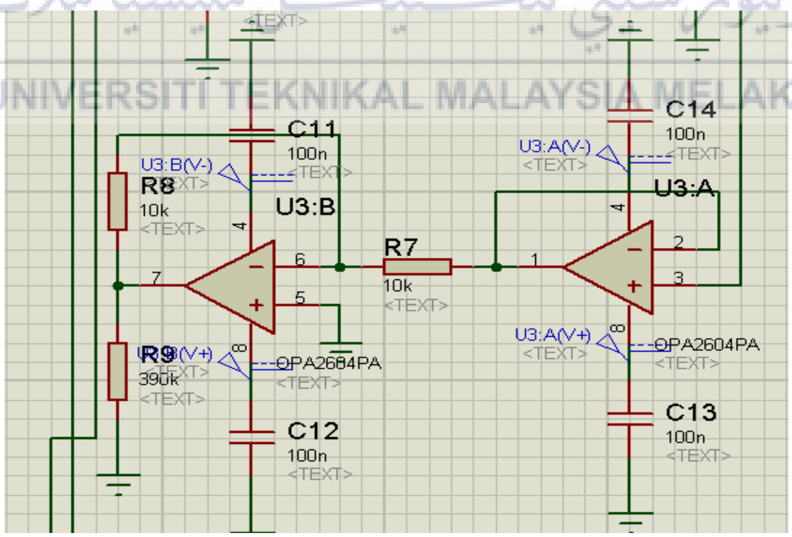
```
hipservo.write(hipangle);
kneeservo.write(kneeangle);
delay (1030);
hipangle = -10 + 97;
kneeangle = 79;//**
hipservo.write(hipangle);
kneeservo.write(kneeangle);
delay (1030);
  hipangle = -1 + 97;
  kneeangle = 80;//**
  hipservo.write(hipangle);
  kneeservo.write(kneeangle);
  delay (1030);
  hipangle = 10 + 97;
  kneeangle = 81;//**
  hipservo.write(hipangle);
  kneeservo.write(kneeangle);
  delay (670);
  hipangle = 21 + 97;
  kneeangle = 79;//**
  hipservo.write(hipangle);
  kneeservo.write(kneeangle);
  delay (670);
```

```
    hipangle = 35 + 97;
    kneeangle = 78;//**
    hipservo.write(hipangle);
    kneeservo.write(kneeangle);
    delay (1442);
    hipangle = 32 + 97;
    kneeangle = 77;//**
    hipservo.write(hipangle);
    kneeservo.write(kneeangle);
    delay (670);
    hipangle = 30 + 97;
    kneeangle = 78;//**
    hipservo.write(hipangle);
    kneeservo.write(kneeangle);
    delay (670);
  }
  else
  {
   hipservo.attach(9,0,0);
   kneeservo.attach(10,0,0);
  }
}
```

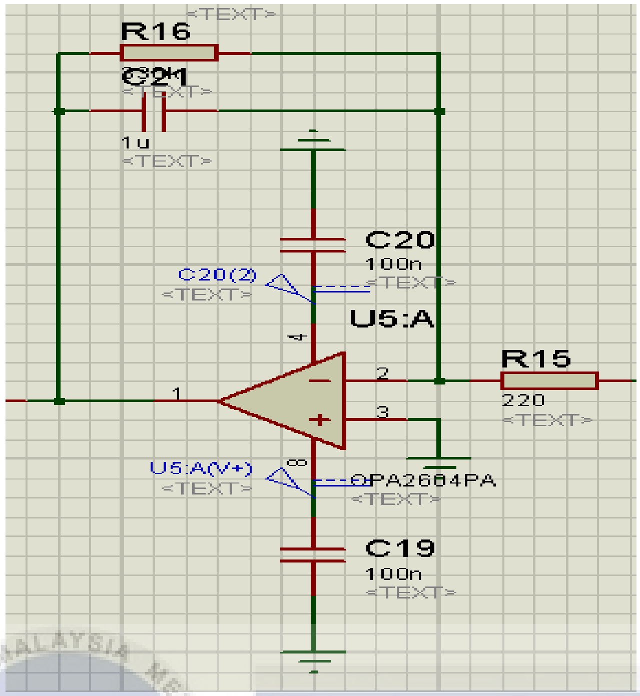
Appendices E: Types of circuit of EMG sensor



Preamplifier circuit



Driven right leg circuit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C19
100n
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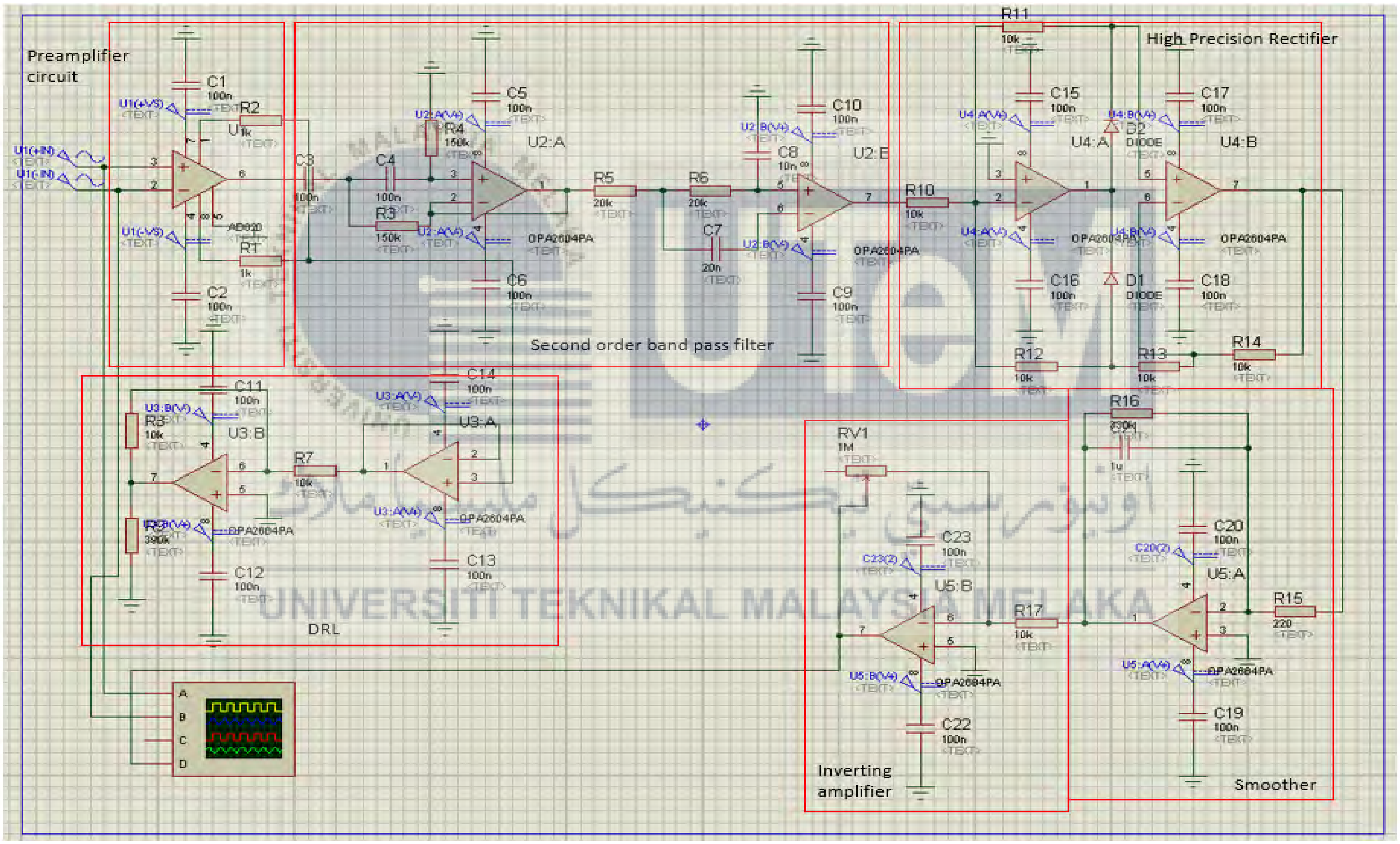
Smoother

Appendices F: Schematic Diagram



Appendices G: Calculated Value for Trajectory Planning

Time (knee)	Time (hip)	Actual time	a0	a1	a2	angle	a0	a1	a2	angle
		0	30	0	0	30	5	1775.1479	-9103.3227	5
		0.01	30	0	0	30	5	1775.1479	-9103.3227	5.168411467
		0.02	30	0	0	30	5	1775.1479	-9103.3227	5.637232578
		0.03	30	0	0	30	5	1775.1479	-9103.3227	6.351843397
		0.04	30	0	0	30	5	1775.1479	-9103.3227	7.257623987
		0.05	30	0	0	30	5	1775.1479	-9103.3227	8.299954413
		0.06	30	0	0	30	5	1775.1479	-9103.3227	9.424214737
		0.07	30	0	0	30	5	1775.1479	-9103.3227	10.57578502
		0.08	30	0	0	30	5	1775.1479	-9103.3227	11.70004534
		0.09	30	0	0	30	5	1775.1479	-9103.3227	12.74237574
		0.1	30	0	0	30	5	1775.1479	-9103.3227	13.6481563
		0.11	30	0	0	30	5	1775.1479	-9103.3227	14.36276708
		0.12	30	0	0	30	5	1775.1479	-9103.3227	14.83158813
	0	0.13	30	-809.58	1401.8702	30	5	1775.1479	-9103.3227	14.99999954
0.01	0.01	0.14	30	-809.58	1401.8702	29.92044387	15	-377.2446	891.8312	14.96316737
0.02	0.02	0.15	30	-809.58	1401.8702	29.68738296	15	-377.2446	891.8312	14.85623681
0.03	0.03	0.16	30	-809.58	1401.8702	29.3092285	15	-377.2446	891.8312	14.6845593
0.04	0.04	0.17	30	-809.58	1401.8702	28.79439169	15	-377.2446	891.8312	14.45348584
0.05	0.05	0.18	30	-809.58	1401.8702	28.15128378	15	-377.2446	891.8312	14.1683674
0.06	0.06	0.19	30	-809.58	1401.8702	27.38831596	15	-377.2446	891.8312	13.83455498
0.07	0.07	0.2	30	-809.58	1401.8702	26.51389948	15	-377.2446	891.8312	13.45739956
0.08	0.08	0.21	30	-809.58	1401.8702	25.53644554	15	-377.2446	891.8312	13.04225213

0.09	0.09	0.22	30	-809.58	1401.8702	24.46436538	15	-377.2446	891.8312	12.59446368
0.1	0.1	0.23	30	-809.58	1401.8702	23.3060702	15	-377.2446	891.8312	12.1193852
0.11	0.11	0.24	30	-809.58	1401.8702	22.06997124	15	-377.2446	891.8312	11.62236767
0.12	0.12	0.25	30	-809.58	1401.8702	20.76447971	15	-377.2446	891.8312	11.10876207
0.13	0.13	0.26	30	-809.58	1401.8702	19.39800683	15	-377.2446	891.8312	10.58391941
0.14	0.14	0.27	30	-809.58	1401.8702	17.97896383	15	-377.2446	891.8312	10.05319065
0.15	0.15	0.28	30	-809.58	1401.8702	16.51576193	15	-377.2446	891.8312	9.5219268
0.16	0.16	0.29	30	-809.58	1401.8702	15.01681234	15	-377.2446	891.8312	8.995478835
0.17	0.17	0.3	30	-809.58	1401.8702	13.49052629	15	-377.2446	891.8312	8.479197746
0.18	0.18	0.31	30	-809.58	1401.8702	11.94531501	15	-377.2446	891.8312	7.978434518
0.19	0.19	0.32	30	-809.58	1401.8702	10.3895897	15	-377.2446	891.8312	7.498540141
0.2	0.2	0.33	30	-809.58	1401.8702	8.8317616	15	-377.2446	891.8312	7.0448656
0.21	0.21	0.34	30	-809.58	1401.8702	7.280241922	15	-377.2446	891.8312	6.622761883
0.22	0.22	0.35	30	-809.58	1401.8702	5.74344189	15	-377.2446	891.8312	6.237579978
0.23	0.23	0.36	30	-809.58	1401.8702	4.229772723	15	-377.2446	891.8312	5.89467087
0.24	0.24	0.37	30	-809.58	1401.8702	2.747645645	15	-377.2446	891.8312	5.599385549
0.25	0.25	0.38	30	-809.58	1401.8702	1.305471875	15	-377.2446	891.8312	5.357075
0.26	0.26	0.39	30	-809.58	1401.8702	-0.088337365	15	-377.2446	891.8312	5.173090211
0.27	0.27	0.4	30	-809.58	1401.8702	-1.425370853	15	-377.2446	891.8312	5.05278217
0.28	0.28	0.41	30	-809.58	1401.8702	-2.69721737	15	-377.2446	891.8312	5.001501862
0.282	0.282	0.412	30	-809.58	1401.8702	-2.943024049	15	-377.2446	891.8312	5.000000016
0.01	0.29	0.42	30	-809.58	1401.8702	-3.895465692	5	2214.7145	-5409.3289	5.216062121
0.02	0.3	0.43	30	-809.58	1401.8702	-5.0117046	5	2214.7145	-5409.3289	5.842611169
0.03	0.31	0.44	30	-809.58	1401.8702	-6.037522872	5	2214.7145	-5409.3289	6.84719117

0.04	0.32	0.45	30	-809.58	1401.8702	-6.964509286	5	2214.7145	-5409.3289	8.19734615
0.05	0.33	0.46	30	-809.58	1401.8702	-7.784252623	5	2214.7145	-5409.3289	9.860620138
0.06	0.34	0.47	30	-809.58	1401.8702	-8.488341659	5	2214.7145	-5409.3289	11.80455716
0.07	0.35	0.48	30	-809.58	1401.8702	-9.068365175	5	2214.7145	-5409.3289	13.99670124
0.08	0.36	0.49	30	-809.58	1401.8702	-9.515911949	5	2214.7145	-5409.3289	16.4045964
0.09	0.37	0.5	30	-809.58	1401.8702	-9.822570759	5	2214.7145	-5409.3289	18.99578668
0.1	0.38	0.51	30	-809.58	1401.8702	-9.979930386	5	2214.7145	-5409.3289	21.7378161
0.11	0.385	0.515	30	-809.58	1401.8702	-9.999994498	5	2214.7145	-5409.3289	24.59822868
0.12	0.01	0.52	-10	929.5139	-1626.0192	-9.908674629	5	2214.7145	-5409.3289	27.54456846
0.13	0.02	0.53	-10	929.5139	-1626.0192	-9.641202594	5	2214.7145	-5409.3289	30.54437946
0.14	0.03	0.54	-10	929.5139	-1626.0192	-9.207340008	5	2214.7145	-5409.3289	33.5652057
0.15	0.04	0.55	-10	929.5139	-1626.0192	-8.616842989	5	2214.7145	-5409.3289	36.57459121
0.16	0.05	0.56	-10	929.5139	-1626.0192	-7.87946765	5	2214.7145	-5409.3289	39.54008003
0.17	0.06	0.57	-10	929.5139	-1626.0192	-7.004970107	5	2214.7145	-5409.3289	42.42921616
0.18	0.07	0.58	-10	929.5139	-1626.0192	-6.003106476	5	2214.7145	-5409.3289	45.20954366

0.19	0.08	0.59	-10	929.5139	-1626.0192	-4.88363287	5	2214.7145	-5409.3289	47.84860652
0.2	0.09	0.6	-10	929.5139	-1626.0192	-3.656305407	5	2214.7145	-5409.3289	50.3139488
0.21	0.1	0.61	-10	929.5139	-1626.0192	-2.3308802	5	2214.7145	-5409.3289	52.57311451
0.22	0.11	0.62	-10	929.5139	-1626.0192	-0.917113365	5	2214.7145	-5409.3289	54.59364767
0.23	0.12	0.63	-10	929.5139	-1626.0192	0.575238982	5	2214.7145	-5409.3289	56.34309232
0.24	0.13	0.64	-10	929.5139	-1626.0192	2.136420728	5	2214.7145	-5409.3289	57.78899249
0.25	0.14	0.65	-10	929.5139	-1626.0192	3.756675755	5	2214.7145	-5409.3289	58.89889219
0.26	0.15	0.66	-10	929.5139	-1626.0192	5.42624795	5	2214.7145	-5409.3289	59.64033545
0.27	0.16	0.67	-10	929.5139	-1626.0192	7.135381197	5	2214.7145	-5409.3289	59.98086631
0.28	0.17	0.68	-10	929.5139	-1626.0192	8.87431938	5	2214.7145	-5409.3289	59.88802879
0.28495	0.17495	0.68495	-10	929.5139	-1626.0192	9.74311976	5	2214.7145	-5409.3289	59.67173491
0.01	0.18	0.69	-10	929.5139	-1626.0192	10.63330639	60	-2327.2973	5579.2816	59.77284955
0.02	0.19	0.7	-10	929.5139	-1626.0192	12.4025861	60	-2327.2973	5579.2816	59.11371533
0.03	0.2	0.71	-10	929.5139	-1626.0192	14.1724024	60	-2327.2973	5579.2816	58.05607303
0.04	0.21	0.72	-10	929.5139	-1626.0192	15.93299918	60	-2327.2973	5579.2816	56.63339834

0.05	0.22	0.73	-10	929.5139	-1626.0192	17.67462032	60	-2327.2973	5579.2816	54.87916695
0.06	0.23	0.74	-10	929.5139	-1626.0192	19.3875097	60	-2327.2973	5579.2816	52.82685455
0.07	0.24	0.75	-10	929.5139	-1626.0192	21.06191122	60	-2327.2973	5579.2816	50.50993682
0.08	0.25	0.76	-10	929.5139	-1626.0192	22.68806875	60	-2327.2973	5579.2816	47.96188946
0.09	0.26	0.77	-10	929.5139	-1626.0192	24.25622618	60	-2327.2973	5579.2816	45.21618816
0.1	0.27	0.78	-10	929.5139	-1626.0192	25.7566274	60	-2327.2973	5579.2816	42.3063086
0.11	0.28	0.79	-10	929.5139	-1626.0192	27.17951628	60	-2327.2973	5579.2816	39.26572648
0.12	0.29	0.8	-10	929.5139	-1626.0192	28.51513672	60	-2327.2973	5579.2816	36.12791748
0.13	0.3	0.81	-10	929.5139	-1626.0192	29.7537326	60	-2327.2973	5579.2816	32.92635731
0.14	0.31	0.82	-10	929.5139	-1626.0192	30.8855478	60	-2327.2973	5579.2816	29.69452163
0.15	0.32	0.83	-10	929.5139	-1626.0192	31.90082621	60	-2327.2973	5579.2816	26.46588615
0.16	0.33	0.84	-10	929.5139	-1626.0192	32.78981172	60	-2327.2973	5579.2816	23.27392655
0.17	0.34	0.85	-10	929.5139	-1626.0192	33.5427482	60	-2327.2973	5579.2816	20.15211853
0.18	0.35	0.86	-10	929.5139	-1626.0192	34.14987955	60	-2327.2973	5579.2816	17.13393777
0.19	0.36	0.87	-10	929.5139	-1626.0192	34.60144964	60	-2327.2973	5579.2816	14.25285996

0.2	0.37	0.88	-10	929.5139	-1626.0192	34.88770237	60	-2327.2973	5579.2816	11.5423608
0.21	0.38	0.89	-10	929.5139	-1626.0192	34.99888162	60	-2327.2973	5579.2816	9.035915968
0.22	0.3861	0.8961	-10	929.5139	-1626.0192	34.97656318	60	-2327.2973	5579.2816	6.767001157
0.23	0.01	0.9	35	-836.6236	4165.415	34.92050306	60	-2327.2973	5579.2816	4.769092057
0.24	0.02	0.91	35	-836.6236	4165.415	34.69867388	60	-2327.2973	5579.2816	3.075664358
0.25	0.03	0.92	35	-836.6236	4165.415	34.35950497	60	-2327.2973	5579.2816	1.72019375
0.26	0.04	0.93	35	-836.6236	4165.415	33.9279888	60	-2327.2973	5579.2816	0.736155922
0.27	0.05	0.94	35	-836.6236	4165.415	33.42911788	60	-2327.2973	5579.2816	0.157026563
0.28	0.06	0.95	35	-836.6236	4165.415	32.88788468	60	-2327.2973	5579.2816	0.016281363
0.29	0.07	0.96	35	-836.6236	4165.415	32.32928171	60	-2327.2973	5579.2816	0.347396012
0.29305	0.07305	0.96305	35	-836.6236	4165.415	32.15927254	60	-2327.2973	5579.2816	0.547410687
0.01	0.08	0.97	35	-836.6236	4165.415	31.77830144	0	3346.4943	-33323.3194	0.301326111
0.02	0.09	0.98	35	-836.6236	4165.415	31.25993638	0	3346.4943	-33323.3194	1.072011165
0.03	0.1	0.99	35	-836.6236	4165.415	30.799179	0	3346.4943	-33323.3194	2.112115246
0.04	0.11	1	35	-836.6236	4165.415	30.42102181	0	3346.4943	-33323.3194	3.221698438

0.05	0.12	1.01	35	-836.6236	4165.415	30.15045728	0	3346.4943	-33323.3194	4.200820825
0.06	0.13	1.02	35	-836.6236	4165.415	30.01247792	0	3346.4943	-33323.3194	4.84954249
0.07	0.14	1.03	35	-836.6236	4165.415	30.0320762	0	3346.4943	-33323.3194	4.967923516

Appendices H: Measurement of the Hip Joint Angle

Hip angle (radian)	Number of times										Average angle	Desired angle
Phases	**1**	**2**	**3**	**4**	**5**	**6**	**7**	**8**	**9**	**10**		
Initial Contact	0.3666	0.3491	0.3840	0.4364	0.3666	0.4538	0.4538	0.6109	0.5062	0.5411	0.4469	0.5237
Loading Response	0.3666	0.3491	0.3840	0.4364	0.3666	0.4538	0.4538	0.6109	0.5062	0.5411	0.4469	0.5237
Mid Stance	0.0873	0.0873	0.1396	0.1571	0.0698	0.1920	0.1746	0.2095	0.2095	0.2269	0.1554	0.2086
Terminal Stance 1	-0.2095	-0.2618	-0.2444	-0.2618	-0.2618	-0.2444	-0.2793	-0.2269	-0.2269	-0.2269	-0.2444	-0.0513
Terminal Stance 2	-0.4015	-0.4713	-0.4538	-0.4364	-0.4713	-0.4538	-0.4713	-0.4189	-0.4364	-0.4189	-0.4434	-0.1746
Pre Swing	-0.2095	-0.2444	-0.2444	-0.2618	-0.2618	-0.2269	-0.2444	-0.1920	-0.2269	-0.2618	-0.2374	-0.0161
Initial Swing 1	0.0000	-0.0349	0.0000	0.0349	-0.0524	0.0000	0.0000	0.0175	-0.0175	-0.0175	-0.0070	0.1700
Initial Swing 2	0.2095	0.1920	0.2095	0.2444	0.1746	0.1571	0.1920	0.2095	0.2095	0.2269	0.2025	0.3676
Mid Swing	0.5935	0.5411	0.5237	0.5237	0.5062	0.5411	0.5237	0.5760	0.5586	0.5935	0.5481	0.6106

Terminal Swing 1	0.5411	0.5062	0.4888	0.4888	0.4713	0.5062	0.4713	0.5237	0.5237	0.5760	0.5097	0.5614
Terminal Swing 2	0.4713	0.4364	0.4364	0.4364	0.4015	0.4713	0.4364	0.4713	0.4713	0.4713	0.4504	0.5242

Appendices I: Measurement of the Knee Joint Angle

Knee angle (radian)	**Number of times**										**Average angle**	**Desired angle**
Phases	**1**	**2**	**3**	**4**	**5**	**6**	**7**	**8**	**9**	**10**		
Initial Contact	0.1396	0.0524	0.0698	0.0524	0.0873	0.1746	0.0175	0.0175	0.0698	0.0175	0.0698	0.0873
Loading Response	0.3317	0.3142	0.2618	0.3666	0.2793	0.3317	0.3142	0.2618	0.2793	0.2618	0.3002	0.2618
Mid Stance	0.2269	0.1746	0.1746	0.1571	0.1746	0.2444	0.1920	0.1571	0.1571	0.1571	0.1815	0.1393
Terminal Stance 1	0.1396	0.1047	0.1047	0.0698	0.1222	0.1571	0.0873	0.0698	0.0873	0.0873	0.1030	0.0873
Terminal Stance 2	0.6109	0.5935	0.6109	0.5935	0.5760	0.6109	0.5586	0.5586	0.5411	0.5411	0.5795	0.4294
Pre Swing	1.3615	1.3790	1.3092	1.3441	1.3266	1.3790	1.3441	1.3441	1.3266	1.3441	1.3458	0.9529
Initial Swing 1	1.3615	1.3790	1.3092	1.3441	1.3266	1.3790	1.3441	1.3441	1.3266	1.3441	1.3458	1.0416
Initial Swing 2	1.2917	1.3092	1.2393	1.2743	1.2743	1.3266	1.2917	1.2917	1.2743	1.2743	1.2847	0.8817
Mid Swing	0.2618	0.2444	0.2095	0.1920	0.2444	0.3142	0.2444	0.2269	0.2269	0.2095	0.2374	0.1182

Terminal Swing 1	0.0873	0.1222	0.0524	0.0175	0.0873	0.1746	0.0873	0.0698	0.0698	0.0524	0.0820	0.0096
Terminal Swing 2	0.1571	0.1571	0.1047	0.0873	0.1396	0.2269	0.1396	0.1222	0.1047	0.1047	0.1344	0.0868

Appendices J: Gantt Chart

FYP Activity	FYP 1 (Week)															FYP 2 (Week)														
	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15
Having Briefing of the Title																														
Literature Review																														
Methodology																														
Preliminary Result																														
FYP1 Report Wrting																														
Lower Limb Rehabilitation Device Hardware Construction																														
EMG Signal Conditioning Circuit Constructed																														
Performing Experiment																														
FYP 2 Report Writing																														

Chapter 1

INTRODUCTION

1.1 Research Background

According to World Health Organization (WHO) [1], stroke is caused by the blood vessel burst due to the clot blocking the blood vessel. This will cut off the supply of the oxygen, nutrients and causing the brain damage. According to The Star: Health [2], there are 2 major type of stroke which are ischaemic and haemorrhagic. Ischaemic stroke occurs when a blood vessel in the brain are blocked. Haemorrhagic stroke occur when a blood vessel inside the brain is weak or burst and causing the blood leak into the brain. The risk of having stroke increase from age to age as shown in the Table 2. From The Star: Health [2], the people after the age of 55 will having the double increase in the risk of having stroke for every decade the person is alive. There are also some other risk that can increase the risk of having stroke like hypertension, diabetes, atrial fibrillation and so on. According to My Malaysia Health [3], stroke is second disease killer in Malaysia as shown in Table 1.1.

Table 1.1: Top disease killer in Malaysia (source: My Malaysia Health) [3]

No	Top 5 Cause of Death	%
1	Coronary heart disease	22.18
2	Stroke	11.67
3	Influenza & pneumonia	9.2
4	Road traffic accidents	7.85
5	HIV/ AIDS	5.53

According to National Stroke Association [4], about 50% of the acute ischemic post stroke patient having a problem which is called hemiparesis. Hemiparesis is the weakness of the muscle of one sided of the body. This will cause the disability of walking, grabbing things and also eating due to the muscle weaken of the one sided of the body and the patient usually will have more than one disability problem.

1.2 Motivation

According to Genentech USA, Inc [5], about 50% of acute ischemic stroke which is one type of stoke will having hemiparesis and 30% of the patient unable to walk without assistance which is the top 1 and top 3 of the disabilities as shown in Figure 1.1. Difficulty of general movement like walking and stabilizing are due to muscle weaken which also called impaired motor control. Impaired motor control of one sided body is called hemiparesis. This disability is the most factor that nowadays most of the post stroke patient are having this disability. Most of the patients will have more than one kind of disability.

The stroke disease in Malaysia is also very serious. It is a second disease killer in Malaysia which is 11.67% as shown in Table 1.1. This means that there are many people in Malaysia having this kind of disease. The post stroke patient in Malaysia also have the disabilities like hemiparesis and unable to walk without assistance.

Nowadays, lower limb rehabilitation device is an important and the only device that can help the post stroke patient who have hemiparesis problem and unable to walk without assistance to regain their walking ability by having training at their lower limb. Therefore, lower limb rehabilitation training for the post stroke patient is needed to allow them to regain the skills and having a normal life like before the stroke.

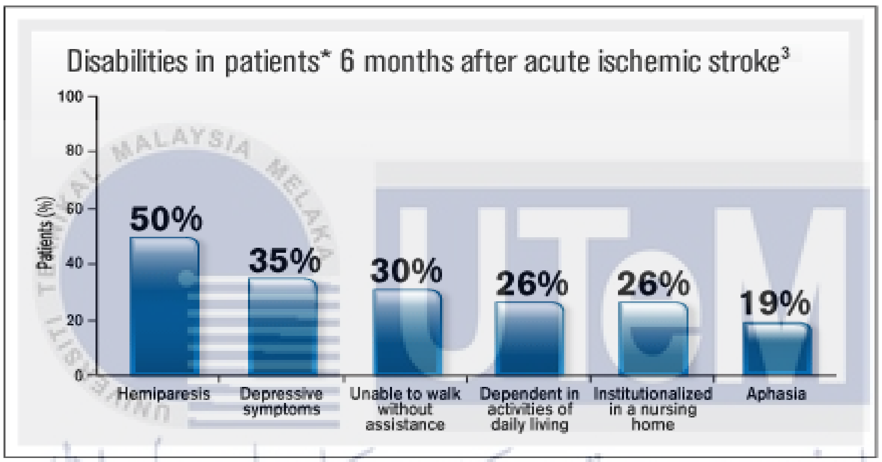


Figure 1.1: Stroke patient disability statistic (source: Genentech USA, Inc) [5]

1.3 Problem Statement

Based on Figure 1.1, hemiparesis and unable to walk without assistance are the most common disabilities faced by post stroke patients who had experienced at least 6 months of acute ischemic. Both disabilities limits or impairs patient’s ability to walk which affect their quality of life and it demands resources for assistive and rehabilitation. Thus, rehabilitation device for lower limb is considered as crucial to restore the walking ability of the post stroke patients. Therefore, rehabilitation device for lower limb is a must for post stroke patient who have this kind of disability to restore their walking ability.

From Iñaki Díaz [6], the lower limb rehabilitation system can be categorized into two types which are wearable and non-wearable. The wearable which is used in both environment, indoor and outdoor, while the non-wearable can be used in indoor only. The non-wearable usually need a treadmill to work to and a therapist as training assistant. It also only allowed the patient to walk in predetermined path. Hence, the wearable type of rehabilitation device is preferred. Even though it might offer complexity in fundamental design, but it have more benefits than the non-wearable rehabilitation system. The wearable rehabilitation system can allow the user to walk their own path rather than just the predetermined path which is very noticeable that almost all systems reviewed have been commercialized. Besides that, the wearable rehabilitation system does not need the treadmill to work. However, rehabilitation training for a wearable rehabilitation system in the market mostly only cover one part of the lower limb only or knee and ankle only or hips, knee and ankle but the training for the hips, knee and ankle is costly due to an extra actuator needed. Moreover, there are don’t have the training for the hips and knee. Therefore, a hips and knee wearable training rehabilitation system is needed for the patient having disability at hips and knee.

According to Iñaki Díaz [6], there are so many type of controller in the lower limb rehabilitation system in the market like Fuzzy logic, PID and so on. Most of the controller are high performance. However, those controllers involves complex algorithm and calculation. Therefore, a simple but with high performance controller is needed for the lower limb wearable rehabilitation system. Therefore to reduce the complexity of the controller, the EMG controller can be implemented in the system as a primary controller to control the lower limb rehabilitation system.

Different type of device using different type of trajectory planning. Different type of trajectory planning could cause the different effect. By using EMG as the controller could cause the consistency of the device decrease but can increase in terms of time response. The cubic polynomial could cause the response time of the device decrease but it will increase the consistency and accuracy of the device. As a summary, a cubic polynomial and EMG based trajectory planning lower limb wearable rehabilitation device with EMG as a primary controller which cover the hips and knee is designed in solving the problems.

1.4 Objective

1. To design a cubic polynomial and EMG based trajectory planning lower limb rehabilitation device for the post stroke patient to regain their walking ability.
2. To analyze overall system performance in terms of accuracy and repeatability.

1.5 Scope

1. In this project, the patient is assumed to have a hemiparesis problem with weaken muscle at thigh and knee but the muscle function normal at the ankle. Therefore, the lower limb rehabilitation prototype is only cover thigh and knee.
2. The test bed used is improved based on crutches developed by previous student, Yong Xian.
3. The prototype has 2 DOF which indicates the rotational movement at hip and knee only.
4. The EMG controller only act as an on off switch to switch on or off the designed rehabilitation device.
5. The trajectory planning of the device is based on the normal forward walking gait of a healthy person.
6. The AgCl electrode is placed at the rectus femoris muscle to detect the EMG of the lower limb and the own signal conditioning is used to condition the signal of muscle.
7. The performance of overall system is analyzed in terms of accuracy and repeatability by analyzing the accuracy of controller signal receiving from the lower limb by comparing them with the muscle sensor and the repeatability prototype is analyzed by comparing the trajectory planning with angle of the device with 10 times of repetition.

1.6 Report Outline

This report is about the development of lower limb rehabilitation device for post stroke patient. In this report, the chapter 1 covers about the research background of the post stroke patient and motivation for designing a lower limb rehabilitation device for post stroke patient. The objective and scope of the design is stated in this chapter. The theory and basic principle and the review of previous related work of the lower limb rehabilitation device is covered in chapter 2 of the report. This report also lists out the methods and techniques used in this design which stated in chapter 3. The result and the conclusion are covered in chapter 4 and chapter 5 in this report.

Chapter 2

LITERATURE REVIEW

2.1 Theory and Basic Principle

Theory and basic principle of the lower limbs rehabilitation device for the post stroke patient is discussed in details in this section. It covers stability of human body, lower limb muscle, electromyography, post stroke patient, weight bearing for the lower limb as well as its motion during walking.

2.1.1 Basic Principle of Lower Limb Rehabilitation Device

2.1.1.1 Post Stroke Patient

In this section, the research will discusses about the disability of the post stroke patients. According to National Institute of Neurological Disorder and Stroke [10], the post stroke patients will have several disability in physical and also have some problem in their mental. The problems like problems of controlling movement, problems of using or understanding the language, sensory disturbance and also emotional disturbance.

According to National Stroke Association [4], about 80% of the post stroke patients will have a problem which is called hemiparesis. Hemiparesis is the one sided muscle weakness especially at the arm and lower limb. This can affect the motion of the arm and also the lower limb like walking or grabbing things.

From National Stroke Association [4], hemiparesis can be separated into 2 parts which are right sided and left sided. The right sided hemiparesis will have the problem of using or understanding the language. The left sided hemiparesis will have memory problems. Damage of the lower brain will cause the motion failure of the arm and also the lower limb which is called ataxia.

2.1.1.2 Motion of the Lower Limb during Walking

According to Aaron M. Dollar [13], there are 8 stages of the motion while a human walking. Figure 4 shows the one gait cycle when a human walking. There are 2 important phase for the gait cycle which is stance phase and swing phase. The stance phase is the phase when the human body moves forward and the swing phase is when the lower limb back into the initial position.

From Aaron M. Dollar [13], human lower limb contains of 7 degree of freedom (DOF) which is the rotation of the ankle, abduction-adduction of the ankle, flexion-extension of the ankle, flexion-extension of the knee, rotation of the thigh, abduction-adduction of the thigh and flexion-extension of the thigh. For a human to walk forward, the lower limb only uses 3 DOF which is flexion-extension of the thigh, flexion-extension of the knee and flexion-extension of the ankle.

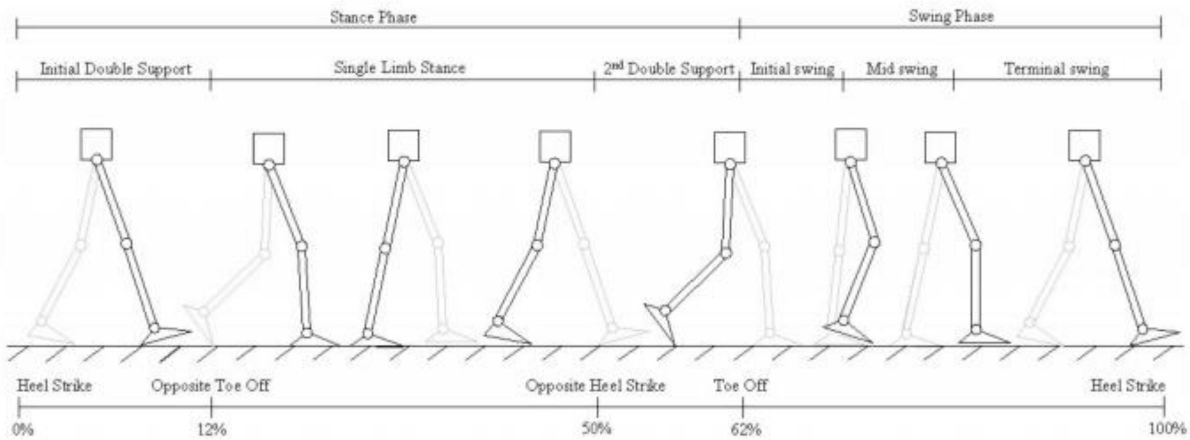


Figure 2.1: Motion of the lower limb during walking (source: Lower Extremity Exoskeletons and Active Orthoses: Challenges and State-of-the-Art) [13]

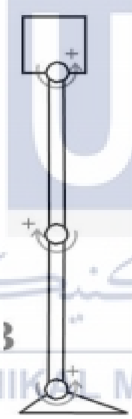


Figure 2.2: DOF of the lower limb (source: Lower Extremity Exoskeletons and Active Orthoses: Challenges and State-of-the-Art) [13]

2.1.1.3 Center of Gravity and Stability

Based on the book, Scientific Basis of Human Motion [7], the center of the gravity of human body is different on height, body size and also as well as the gender. The standing position will affect the stability of the human body. Female center of gravity is 55% of the body height and male is 57% of the body height.

According to Nancy Hamilton et. al. [7], the stability of the human body can be increased with the help of some instrument like using a crutch while walking or standing. The crutch can increases the base support area of the human body. With the increase of the base support area, the stability of the human body will also increases.

Figure 2.3: Center of gravity of human body (source: *Kinesiology : Scientific Basis of Human Motion)* [7]

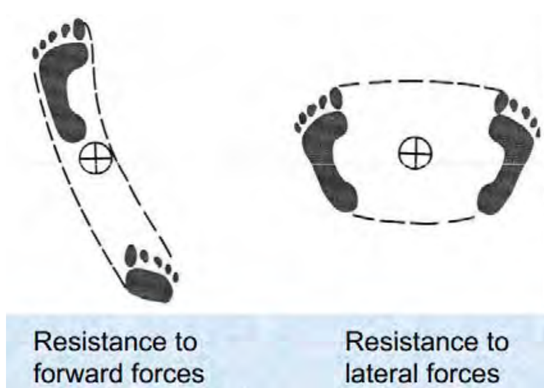


Figure 2.4: Shape of the base support of the human body based on the standing position (source: *Kinesiology : Scientific Basis of Human Motion)* [7]

2.1.1.4 Weight Bearing of the Lower Limb

The average of the human lower limb is 8kg for the person that weight 51kg. For the hemiparesis patient, according to Ruth Dickstein [11], the hindfoot can stands for 39% of the body weight of the post stroke patient and 50% for forefoot before the rehabilitation of the lower limb treatment. After the treatment, the hind foot can stands for 47% of the body weight which increase 8% but the forefoot still remain the same which is 50% of the body weight.

2.1.1.5 Human Height

According to Plagenhoef, S., Evans [12], percentage of the length of thigh is 24.05% of the overall body height and knee length is 25.2% of the overall body height. This means that for group of persons who are about 160cm to 180cm, their length of thigh and leg are 38cm to 43cm and 40cm to 45cm.

2.1.2 Basic Principle of EMG Sensor

2.1.2.1 Electromyography (EMG)

According to Konrad, P [9], EMG is a myoelectric signal which produced by the muscle fiber membranes. The EMG in nowadays is widely used in many fields especially in rehabilitation for training purpose. It can be detected by using the muscle sensor.

According to Konrad, P [9], there are few types of electromyography like Surface Electromyography, Fine Wire Electromyography, Neuromuscular Electrical stimulation and EMG- Triggered stimulation. Most of the application is using the Surface Electromyography which detect the muscle signal by attaching the sensor on the surface of the muscle due to its easy to use and easy to get the signal from the muscle.

According to Konrad, P [9], the EMG signal processing contains of three parts which are signal conditioning, normalization and classification. The signal conditioning consists of rectifier, amplifier, bandpass filter and smoothing the signal. The normalization consists of amplitude normalization, time normalization and related parameters. The classification is the on or off of the muscle of the human body either in lower limb or upper limb.

From Konrad, P [9], the accuracy of the EMG is tested by comparing the shape of the graph of the rectified EMG signal with the EMG controller prototype. The EMG controller is tested using the same muscle and performing the same activities during the test and it also needs to depend on which part of the body need to be tested. The EMG controller needs to be test for a number of times to ensure the accuracy of the EMG when comparing with the rectified EMG signal.

2.1.2.2 Lower Limb Muscle

From He. H. et. al. [8], the lower limb comprises of many muscles that assist the body movement as well as daily activities that involve the lower limb motion. Flexion is the motion of the leg that moves forward and extension is the motion of the leg that moves backward. Adduction is the motion of the leg that moves outwards and abduction is the motion of the leg that moves inwards.

There are some of the muscles that help the motion of the thigh and leg:

- Iliacus – for flexion of the thigh.
- Psoas major – for flexion of the thigh
- Tensor fasciae latae – flexion and abduction of the thigh

- Adductor magnus – for flexion and adduction of the thigh
- Gluteus maximus – for extension of the thigh
- Rectus femoris – for leg extension
- Vastus lateralis – for leg extension
- Semitendinosus – for leg flexion
- Semimembranosus – for leg flexion

Figure 2.5: Muscle of the lower limb (source: A Study on Lower-Limb Muscle Activities during Daily Lower-Limb Motions) [8]

2.2 Review of Previous of Related Works

In this section, the research is about the lower limb robotics rehabilitation system. This section will discuss about the types of the robotic rehabilitation system for the lower limb that existing in the market and also the few design of lower limb rehabilitation system.

2.2.1 Types of Lower Limb Robotics Rehabilitation System

According to Iñaki Díaz [6], nowadays, there are many kind of lower limb robotics rehabilitation system in the market. There are 2 types of the rehabilitation system which is non-wearable types as shown in Figure 2.6 and wearable types as shown in Figure 2.7. The non-wearable type rehabilitation system that contain a multifunction big machine which record the reading of the training and improve the training but need the therapist to assist and required a treadmill as shown in the Figure 2.6. The wearable type rehabilitation system is the system can use at the outdoor and also can use it without the help of a therapist but only contain one function. However, most of the wearable type rehabilitation system is still in trial stage and most of them is for the knee and ankle training.

According to Bortole, M. [14], the type of the design is wearable lower limb rehabilitation system which is covered thigh, knee and ankle and used 6 DOF as shown in Figure 2.8. The lower limb rehabilitation system design by Enguo, C. [15] is non-wearable lower limb rehabilitation system which is only covered thigh and knee and used 2 DOF as shown in Figure 2.9. Same as Enguo, C, the lower limb rehabilitation system design by Koceska, N. [16] is non-wearable lower limb rehabilitation system but cover thigh, knee and ankle and it uses 10 DOF as shown in Figure 2.10. The Figure 2.11 shows the design which is a non-wearable lower limb rehabilitation system designed by Kari A. Danek [17]. The design only covers the ankle and it is 1 DOF.

According to Erhan Akdogˇan [18] as shown in the 2.12, the lower limb rehabilitation system is a non-wearable and 3 DOF device which covers thigh and knee only. The lower limb rehabilitation system design by Yixiong Chen [19] is non-wearable but it covers thigh, knee and ankle and it is a 3 DOF device as shown in the Figure 2.13. Figure 2.14 (Nelson Costa [20]) shows the design which is a wearable and 10 DOF lower limb rehabilitation system. It covers thigh, knee and ankle. Figure 2.15 (Carberry et al [21]) and Figure 2.16 (Park et al. [21]) shows the designs which are wearable and they are 2 DOF lower limb rehabilitation device that cover ankle only. Active modular elastomer for soft wearable assistance robots (Park et al. [21]) as shown in the Figure 2.17 is also a 2 DOF and wearable lower limb rehabilitation device but covers knee only.

Figure 2.6: Non-wearable type rehabilitation system [7]

Figure 2.7: Wearable type rehabilitation system [7]

Figure 2.8: Lower limb rehabilitation system designed by Bortole, M. [14]

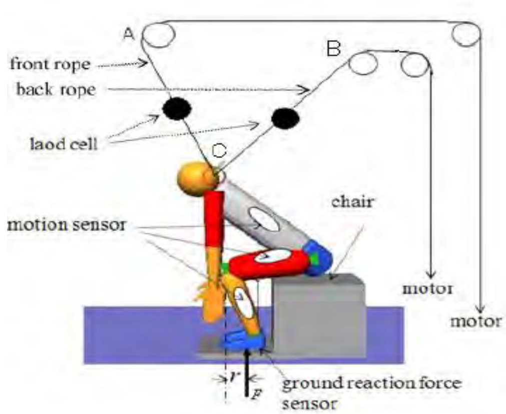


Figure 2.9: Lower limb rehabilitation system designed by Enguo, C [15]

Figure 2.10: Lower limb rehabilitation system designed by Koceska, N [16]

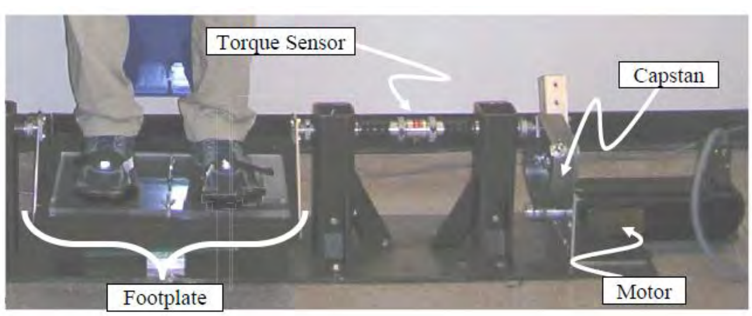


Figure 2.11: Lower limb rehabilitation system designed by Kari A. Danek [17]

Figure 2.12: Lower limb rehabilitation system designed by Erhan Akdogˇan [18]

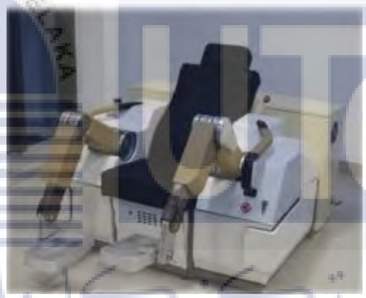


Figure 2.13: Lower limb rehabilitation system designed by Yixiong Chen [19]

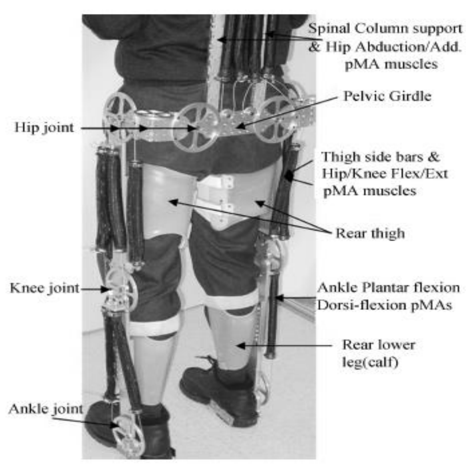


Figure 2.14: Lower limb rehabilitation system designed by Nelson Costa [20]

Figure 2.15: Active ankle-foot orthosis (AAFO) designed by Carberry et al. (source: Recent Trends in Lower-Limb Robotic Rehabilitation Orthosis: Control Scheme and Strategy for Pneumatic Muscle Actuated Gait Trainers) [21]

Figure 2.16: Bio-inspired active soft orthotic device for ankle foot pathology designed by Park et al. (source: Recent Trends in Lower-Limb Robotic Rehabilitation Orthosis: Control Scheme and Strategy for Pneumatic Muscle Actuated Gait Trainers) [21]

Figure 2.17: Active modular elastomer for soft wearable assistance robots designed by Park et al. (source: Recent Trends in Lower-Limb Robotic Rehabilitation Orthosis: Control Scheme and Strategy for Pneumatic Muscle Actuated Gait Trainers) [21]

2.2.2 Trajectory Planning

According to Magdo Bortole [14] and Nelson Costa [20], only both of them using the cubic polynomial as the trajectory planning while Enguo, C. [15], Koceska, N. [16], Yixiong Chen [19] and Erhan Akdogˇan [18] are using the inverse kinematics as the trajectory planning. The cubic polynomial trajectory planning in their design are used to control the movement of lower limb rehabilitation device based on the desired angle either sit and stand angle or walking motion angle. However, the inverse kinematics trajectory planning in their design are used to control the movement of lower limb rehabilitation device based on the force of the lower limb to control the torque of the actuator.

2.2.3 Controller

In Enguo, C. [15] design, the controller used is EMG which using the EMG signal and convert it into forces to calculate the muscle forces and the position of the actuator will be controlled based on the muscle forces. According to Bortole, M. [14], Koceska, N. [16], Kari A. Danek [17] and Nelson Costa [20], the designs are using the Fuzzy Logic and PID as the controller to control the actuator and also the position of the lower limb rehabilitation system. Both of them involve many calculation. In Erhan Akdogˇan [18] and Yixiong Chen [19], they are using the impedance as the controller to control the lower limb degree of movement. This type of controller has algorithm that contain calculation. Active ankle-foot orthosis (AAFO) designed by Carberry et al. [21] using the feedback control that utilizes a fuzzy logic gait phase detection system as the controller while bio-inspired active soft orthotic device for ankle foot pathology designed by Park et al. [21] using the feed-forward and feedback controllers as the controller of the device. Both of this controller is used to control the ankle joint angle. Active modular elastomer for soft wearable assistance robots designed by Park et al. [21] control the motion of the knee through shape and rigidity control.

2.2.4 Sensor

In Enguo, C. [15] design, he use the motion sensor to sense the motion of the rehabilitation system and hard type sensor to receive the signal from the lower limb muscle. The main sensor used in Bortole, M. [14], Koceska, N. [16] and Nelson Costa [20] are potentiometer to detect the motion of the device. According to Kari A. Danek [17], Erhan Akdogˇan[18] and Yixiong Chen [19] designs, they use the torque sensor as their main sensor to detect the force between the lower limb and the device. Active ankle-foot orthosis (AAFO) designed by Carberry et al. [21] using the force sensory resistor to identify contact between foot and ground and rotary encoder to measure the angular displacement of ankle joint. Bio-inspired active soft orthotic device for ankle foot pathology designed by Park et al. [21] and active modular elastomer for soft wearable assistance robots designed by Park et al. [21] using the strain sensor as the primary sensor to measure the angle of ankle joint.

2.2.5 Performance Analysis from Previous Related Work

According to Bortole, M. [14], the performance analysis of this design is using a high speed infrared camera. The lower limb is attached to a marker that can reflect the infrared. This test can used to detect the angle of the lower limb. In Enguo, C. [15] design, the performance of the design was analyzed by using a video camera in measuring the movement of the lower limb. This method is using the repeatability test to test the performance by using the design testing on few person. Based on Koceska, N. [16] design, this design using a video camera, Panasonic Nv-gs400eg with 4.0 MPixel, 3CCD, the Leica Dicomar objective and a focal distance of 3.3mm to analyze the repeatability of the design. The camera is set 13m from the treadmill and the lower limb is marked with three marker. Then the distance of the lower limb movement is measured using the pixel of the camera by transforming it into measuring units. The results from the experiments, at a confidence level $p=0.05$, have shown that the maximum absolute error is 3.5mm, which has no significant impact on the rehabilitation process and can be treated as an acceptable result. According to John J. Craig [22], the design can be analyzed by comparing the trajectory planning with the angle of the robot. According to Sharif Muhammad Taslim Reza [23], the EMG signal is perpendicular to the angle of the lower limb movement. Therefore, the EMG signal can be compare with the angle and use the value for programming the device.

2.3 Summary

Based on the Table 2.1, there are 7 example of design. There are 5 non-wearable and 5 wearable type of lower limb rehabilitation system. According to the Bortole, M. [14] and Nelson Costa [20] designs, their designs are wearable which are more preferable from the user compare to the non-wearable designs (Iñaki Díaz [6]). There are 4 designs are using the inverse kinematics as the trajectory planning and 2 designs are using the cubic polynomial as the trajectory planning. The cubic polynomial trajectory planning is used to control the movement of lower limb rehabilitation device based on the desired angle either sit and stand angle or walking motion angle but the inverse kinematics trajectory planning in their design are used to control the movement of lower limb rehabilitation device based on the force of the lower limb to control the torque of the actuator. Since the post stroke patient is assume to have weaken muscle at thigh and knee, therefore it is suitable to use cubic polynomial as the trajectory planning rather than inverse kinematics because the inverse kinematics requires the patient to have force at the lower limb to move device.

Different sensor is used in each of the design. However, the performance of all the sensor used is almost the same but costly due to its high accuracy. Therefore, the design of the lower limb rehabilitation system can use the own signal conditioning as EMG sensor so that to reduce the cost but remain the performance.

The device controller is used as a switch which switch on and off the lower limb rehabilitation device. In order to have that kind of controller, a simple controller like EMG controller can be implemented into the device as a switch. Moreover, PID, Fuzzy Logic and impedance controller have more calculation and algorithm than the EMG controller which is used by Enguo, C. [15]. Therefore, in order for the controller used as an on off switch in the device, the EMG controller which do not have complex calculation and algorithm like others controller is the best choice among them.

Based on Table 2.1, there are no wearable lower limb rehabilitation system which cover thigh and knee. Therefore, a 2 DOF wearable cubic polynomial based trajectory planning lower limb rehabilitation device which covers the thigh and knee with EMG as the controller and own signal conditioning as the EMG sensor is needed for the post stroke patient to regain their walking ability.

From the research of the literature review, the best performance analyzing of the lower limb rehabilitation system design is repeatability analysis and accuracy analysis. The prototype can be analyzed either in angle of the lower limb or distance of the lower limb movement as stated in the in section 2.2.5. However, those methods need the video camera to help measuring. This is very costly due to the video camera. However, the performance of the prototype can be analyzed by measuring the angle of the lower limb rehabilitation system with trajectory planning. The EMG sensor is analyzed by comparing the peak to peak voltage value of the muscle sensor but with the condition of attached it with the same muscle and performing the same activities during the test. The value can be obtained by using the oscilloscope. The EMG sensor need to be measured for at least 3 times to ensure the accuracy of the EMG when comparing with the muscle sensor.

Table 2.1: List of detail of the lower limb rehabilitation system

Title	Author	Type	Position of lower limb	DOF	Trajectory Planning	Sensor	Controller
Design and Control of a Robotic Exoskeleton for Gait Rehabilitation	Bortole, M.	Wearable	Thight, knee and ankle	6	Cubic Polynomial	Potentiometer and strain gauge	PID and Fuzzy Logic
Human Lower Limb Dynamic Analysis Using Wearable Sensors and Its Applications to a Rehabilitation Robot	Enguo, C.	Non-wearable	Thight and knee	2	Inverse Kinematics	Motion Sensor and hard type sensor	EMG
Control Architecture of a 10 DOF Lower Limbs Exoskeleton for Gait Rehabilitation	Koceska, N.	Non-wearable	Thight, knee and ankle	10	Forward and Inverse Kinematics	Potentiometer	Fuzzy Logic
Lower Limb Motor Coordination and Rehabilitation Facilitated through Self-Assist	Kari A. Danek	Non-wearable	Ankle	1	NA	Torque sensor	PID controller

The design and control of a therapeutic exercise robot for lower limb rehabilitation: Physiotherabot	Erhan Akdog˘an	Non-wearable	Thigh and knee	3	Inverse Kinematics	Torque sensor	Impedance controller
The FES-assisted control for a lower limb rehabilitation robot: simulation and experiment	Yixiong Chen	Non-wearable	Thigh, knee and ankle	3	Inverse Kinematics	Torque and angle sensor, EMG sensor	Impedance controller
Joint Motion Control of a Powered Lower Limb Orthosis for Rehabilitation	Nelson Costa	Wearable	Thigh, knee and ankle	10	Cubic Polynomial	Potentiometer	PID controller
Recent Trends in Lower-Limb Robotic Rehabilitation Orthosis: Control Scheme and Strategy for Pneumatic Muscle Actuated Gait Trainers	Yamamoto, S Designed by: Carberry et al.	Wearable	Ankle	2	NA	Force sensory resistor and rotary encoder	Feedback control that utilizes a fuzzy logic gait phase detection system

Recent Trends in Lower-Limb Robotic Rehabilitation Orthosis: Control Scheme and Strategy for Pneumatic Muscle Actuated Gait Trainers	Yamamoto, S Designed by: Park et al.	Wearable	Ankle	2	NA	Strain sensor	Feed-forward and feedback controllers
Recent Trends in Lower-Limb Robotic Rehabilitation Orthosis: Control Scheme and Strategy for Pneumatic Muscle Actuated Gait Trainers	Yamamoto, S Designed by: Park et al.	Wearable	Knee	2	NA	Strain sensor	Through shape and rigidity control

Chapter 3

RESEARCH METHODOLOGY

This chapter is about methods and techniques which are used in this lower limb rehabilitation device to achieve the objectives. The detail of the experiment set up will also be discussed in this section.

Figure 3.1 shows the overall methodology flow chart of this research. Firstly, literature review is conducted to perform gap analysis. Based on that, motivation, problem statement and objectives of this research are conducted. After that, the lower limb rehabilitation system is designed and tested. If the device have problem, then the troubleshooting is carried out. The design of lower limb rehabilitation system is discussed in section 3.1. In this section, device design, material selection, trajectory planning and software development are included. Then, EMG signal processing which includes the schematic circuit is conducted at section 3.2. If the circuit have problem, then troubleshooting is carried out to solve the problem. Finally, overall design performance analysis which contains of the performance analysis procedure is conducted at section 3.3.

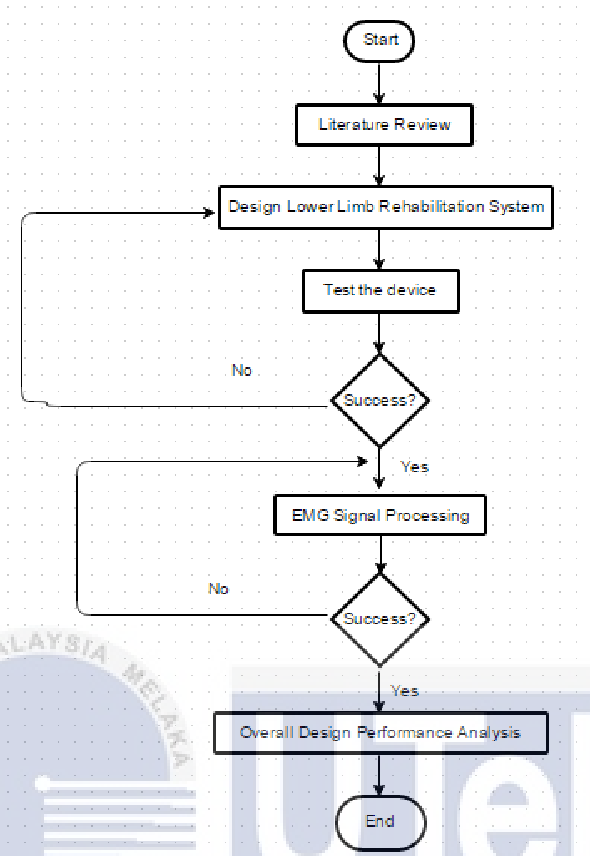


Figure 3.1 Overall methodology flow chart

3.1 Designing the Lower Limb Rehabilitation System for Post Stroke Patient

3.1.1 Overall System

Figure 3.2 shows block diagram of lower limb rehabilitation system. The EMG controller is used as the controller for mapping the EMG signal, servo motor is used as actuator for the motion of system and EMG signal conditioning is used for condition the EMG signal which receives from AgCl electrode.

Material used for the system is Aluminum Alloy due to its light weight and hard material characteristic. EMG signal conditioning is a self-made EMG sensor which will be discuss in section 3.2. The servo motor is Sonxun SXSV50 12 - 24 V High Power Servo Motor as shown in Appendices A. This motor is a type of DC geared motor which convert the DC motor to servo motor. Microcontroller used is Arduino UNO board as shown in Appendices A due to its short and easy understanding coding. This microcontroller is used as a medium for the EMG controller.

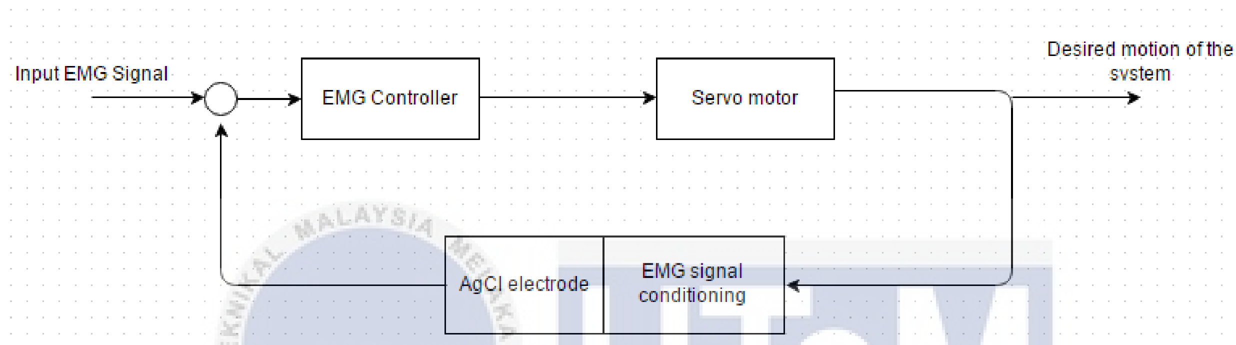


Figure 3.2: Block diagram of lower limb rehabilitation system

3.1.2 Trajectory Planning

Lower limb rehabilitation system is designed using 2 DOF which covers thigh and knee. The leg is tied with the design using ropes. Figure 3.3 shows the free body diagram of the design. The hip joint is fixed which means that when hip joint rotate, the parts below hip joint will move and hip joint will not move. Servo motors is placed at the hip and knee joint which can rotate in terms of angle.

The purpose of the trajectory planning is to find the desired angle of the device so that the device can perform with the desired angle. The desired angle is based on the degree of motion of human walking.

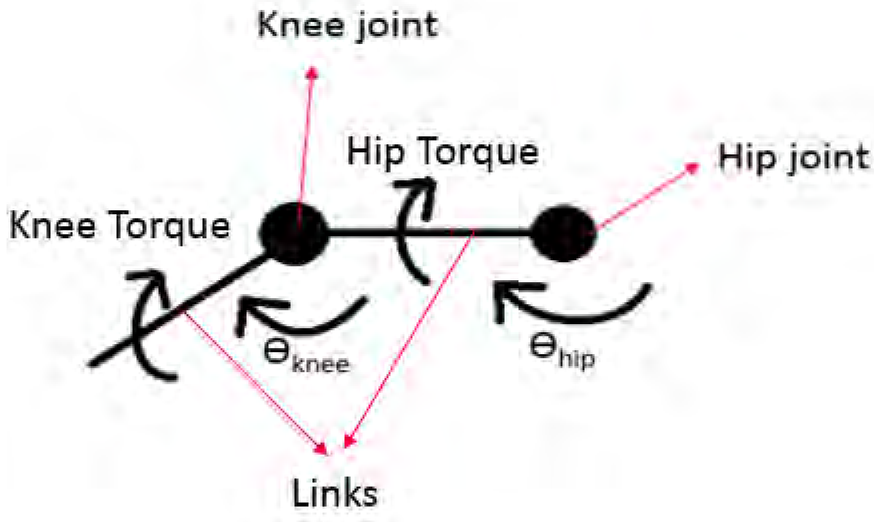


Figure 3.3: Free body diagram of the lower limb rehabilitation system

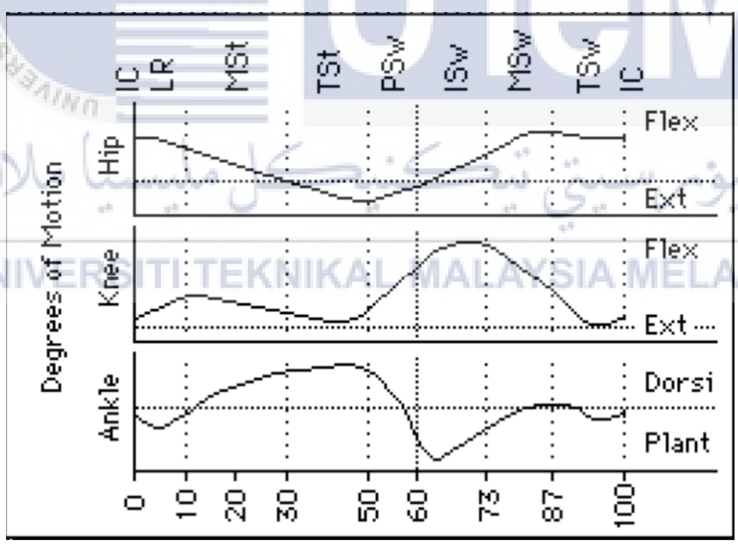


Figure 3.4: Degree of motion of the lower limb during walking (source: Rancho Los Amigos National Rehabilitation Center) [24]

Table 3.1: Degree of motion based on the hip and knee

	Degree of Motion (°)							
Joint\Phase	IC	LR	MST	TST	PSW	ISW	MSW	TSW
Hip	30	30	5	-10	0	25	35	30
Knee	5	15	8	5 then 12	40	60 then 55	20	0 then 5

According to Rancho Los Amigos National Rehabilitation Center [24], the degree of motion of human walking is shown in Figure 3.4. In this design, the hip joint and knee joint degree of motion are taken. Table 3.1 show the degree of motion of the hip and knee which will be taken as the reference to calculate the desired velocity of the device.

From John J.Craig [22],

The initial degree, θ of the slope of knee and hip joint changes sign at the via point,

$$\theta(t_{i-1}) = \theta_{i-1}, where\ i = 1,2,3, ... \tag{3.1}$$

The final degree, θ of the slope of knee and hip joint changes sign at the via point,

$$\theta(t_i) = \theta_{i,}\ where\ i = 1,2,3, ... \tag{3.2}$$

The initial and final velocity,

$$\frac{\delta\theta_{i-1}}{\delta t_{i-1}} = \frac{\delta\theta_i}{\delta t_i} = 0 \tag{3.3}$$

$$joint\ angle, \theta(t) = a_0 + a_2t^2 + a_3t^3 \tag{3.4}$$

$$angular\ velocity, \frac{\delta\theta(t)}{\delta t} = 2a_2t + 3a_3t^2 \tag{3.5}$$

$$a_0 = \theta_{i-1} \tag{3.6}$$

$$a_2 = \frac{3}{t^2}(\theta_i - \theta_{i-1}) \tag{3.7}$$

$$a_3 = -\frac{2}{t^3}(\theta_i - \theta_{i-1}) \tag{3.8}$$

3.1.3 Mechanical Design

The mechanical design consists of 2 main parts which are hip joint and knee joint which is designed by previous student, Yong Xian. The rehabilitation device that has been designed by using Solidworks comprises of few components as shown in Appendices B and Appendices C. Figure 3.5 shows a complete hardware of the device which consists of EMG circuit, Arduino UNO board, wooden block which use to hang the device, hip joint and knee joint.

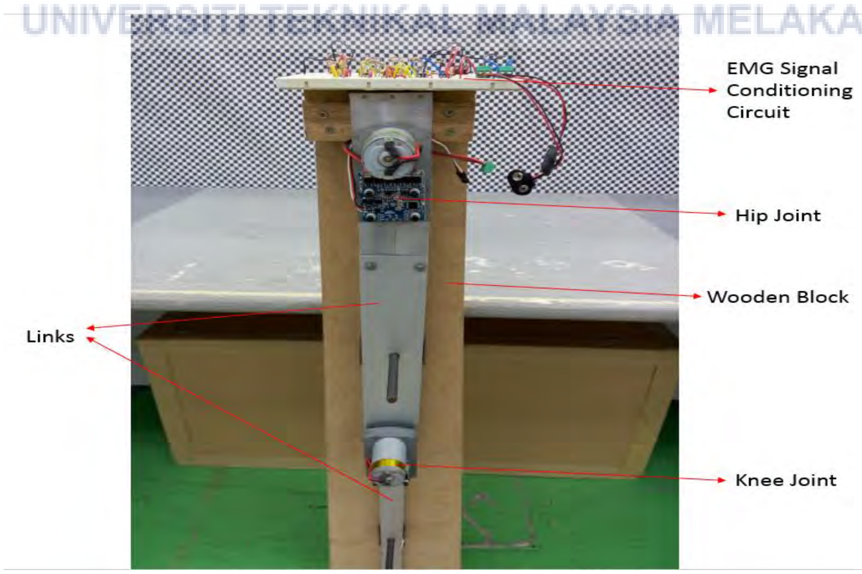


Figure 3.5: Complete Hardware of the Device

3.1.4 Bill of Component

Table 3.2 shows the quantity of components are used in the lower limb rehabilitation device. The quantity of Sonxun SXSV50 12 - 24 V High Power Servo Motor is 2 due to the motor is placed at hip and knee joint. This device also requires 1 Arduino UNO board to control the servo motor and receive the EMG signal from the EMG sensor. The EMG sensor is self-made which means that the EMG sensor no need to buy for the device.

Table 3.2 Bill of component

Component	Quantity
Sonxun SXSV50 12 - 24 V High Power Servo Motor	2
Arduino UNO board	1
EMG sensor	Self-made

3.1.5 Software Development

3.1.5.1 Software Design

Figure 3.6 shows the flow chart of the program of lower limb rehabilitation system. First, the EMG signal from the lower limb muscle need to be detected then only the servo motor rotate to the desired angle. The servo motor is rotated based on the trajectory planning. The servo motor stop after 1 cycle if there is no input EMG signal. If emergency stop button is pushed, during the servo motor rotate or before the servo motor rotate, the motor stop. Appendices D shows the program coding of the device.

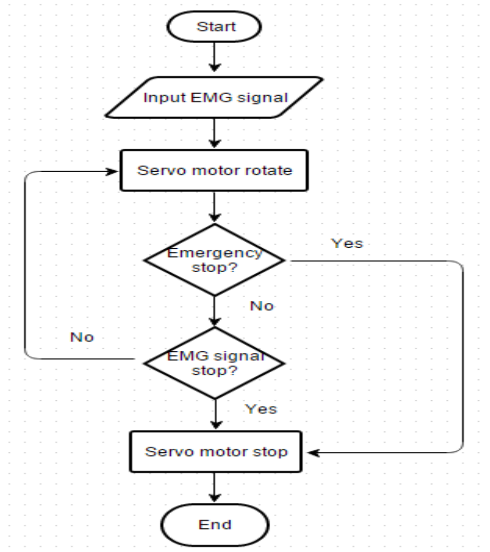


Figure 3.6: Flow chart of the program of lower limb rehabilitation system

3.1.5.2 Electric Circuit Design

Figure 3.7 shows the circuit design of the system which using the Arduino UNO board as a medium to connect the EMG signal conditioning, emergency switch and servo motor. EMG signal conditioning is connected at pin 0 and emergency switch is connected at pin 1. Both of the pin is used for analog and digital input.

Hip servo motor board is connect at pin 9 and knee servo motor board is connected at pin 10. Both of the pin is used for PWM connection. The servo motor board is already connected the servo motor when buy. The voltage source for the servo motor board is 24V and the Arduino board is 5v.

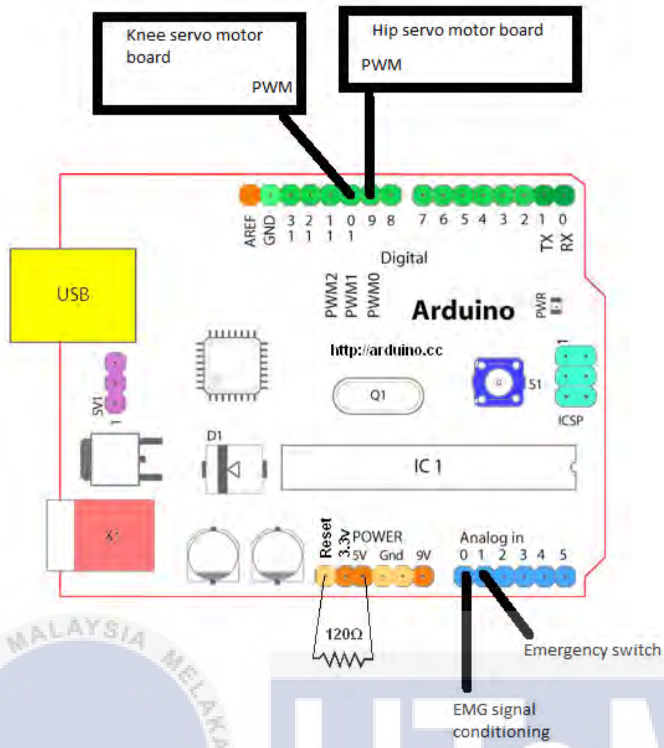


Figure 3.7: Circuit design of the system

3.2 EMG Sensor

3.2.1 Block Diagram of the EMG Sensor Circuit

Figure 3.8 shows the block diagram of the EMG sensor circuit. The sensor is used to detect the muscle signal. The signal conditioning is used to condition the signal so that the output signal is desired. The AgCl electrodes is used as the electrode to detect the signal from the muscle. In the signal conditioning stage, there are preamplifier circuit, filter circuit, noise cancellation circuit, inverting amplifier circuit, rectifier and smoothing circuit. After that, the output signal from the signal conditioning stage is displayed on the oscilloscope to ensure the output signal is desired.

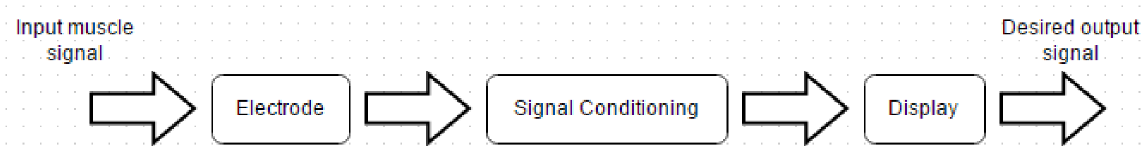


Figure 3.8: Block diagram of the EMG signal conditioning circuit

3.2.2 EMG Sensor Schematic Circuit

3.2.2.1 Type of Circuit of EMG Sensor

The schematic circuit is made using the Proteus software. The circuit contains of preamplifier circuit, driven right leg circuit, second order band pass filter, inverting amplifier, rectifier and a smoother. The AgCl electrode is used to detect the muscle signal from the lower limb.

The preamplifier circuit as shown in Appendices E is used to differentiate the input EMG signal. The capacitor is used to ensure the supply voltage will not drop. This Driven Right Leg as shown in Appendices E is used to reduce the noise produce by the body.

Frequency of the lower limb is 10Hz to 500Hz (Konrad, P [9]). To get the 10Hz and 500Hz range of frequency, a band pass filter as show in Figure 3.9 is used. The band pass filter consists of high pass filter and low pass filter.

For the high pass filter, the capacitor value is fixed at 100nF.

$$R = 1/(2\pi \ x \ f \ x \ C) \tag{3.9}$$

For low pass filter, the capacitor value is fixed at 10n.

$$R = 1/(2\pi \ x \ f \ x \ C) \tag{3.10}$$

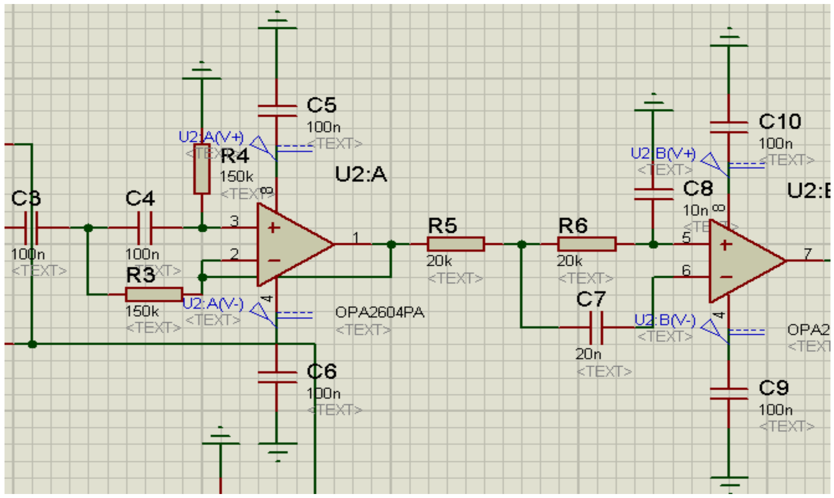


Figure 3.9: Band pass filter

The inverting amplifier as shown in Figure 3.10 is used to invert input signal.

$$Gain = -\left|\frac{R10}{R11}\right|, \text{ where gain} = 1 \tag{3.11}$$

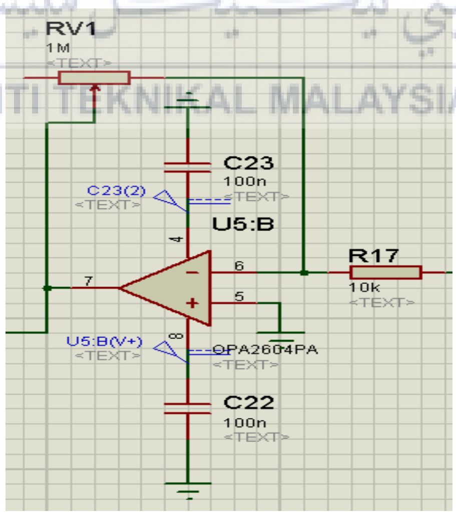


Figure 3.10: Inverting amplifier

The rectifier used in the circuit is high precision rectifier as shown in Figure 3.11 is to rectify the circuit.

According to Ken Bigelow [25], the resistor value can be calculated using the equation,

$$R13 = R19 = R18 \tag{3.12}$$

$$R14 = R12 = 2R13 \tag{3.13}$$

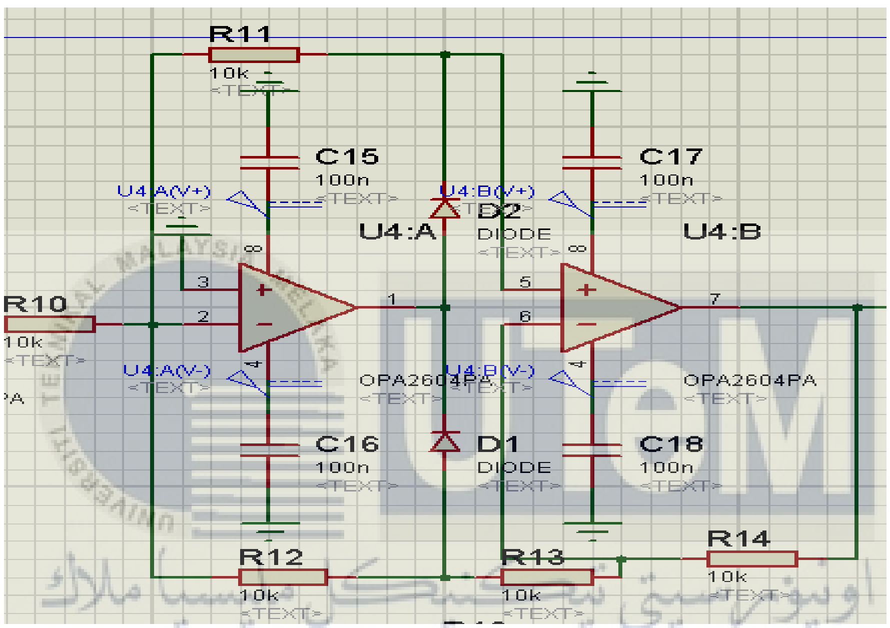


Figure 3.11: Rectifier

The smoother as shown in Appendices E is used to smooth the input signal so that the signal can become dc voltage. By combining the smoother and rectifier, the raw EMG signal become linear envelop. The purpose of these 2 circuit is to make the EMG signal become linear envelop so that feature extraction from the EMG signal can be done.

3.2.2.2 EMG Sensor Complete Circuit

Appendices F shows the complete circuit of the EMG sensor. The inputs are connected to the AgCl electrode which used to receive the EMG signal. The output will be connected to the oscilloscope to see the trend of the signal. Table 3.3 shows the resistor and capacitor value. The resistor value for high pass filter is calculated using equation 3.9 by fixing the capacitor value at 100nF. The resistor value of low pass filter is calculated using equation 3.10 by fixing the capacitor value at 10nF while the resistor value for inverting amplifier is calculated using equation 3.11 by fixing one of the resistor value at 10kΩ. The resistor, R13 of rectifier is fixed at 20kΩ and using the equation 3.12 and 3.13 to calculate the rest resistor value.

Table 3.3: Resistor value and capacitor value for amplifier

Amplifier	Resistor value (Ω)	Capacitor value (F)
High pass filter	150k	100n
Low pass filter	32k	10n
Inverting amplifier	10k	-
Rectifier	R13=R18=R19=20k R12=R14=40k	-

3.2.2.3 Simulation of EMG Sensor

After the schematic circuit diagram is made, the circuit is then simulated using Proteus which is as shown in Figure 3.12. The red and purple line is the sinusoidal input. The black line is the reference which show the zero voltage value and the purple line is the output of the circuit which is a dc voltage which the input signal is rectified and smoothed.

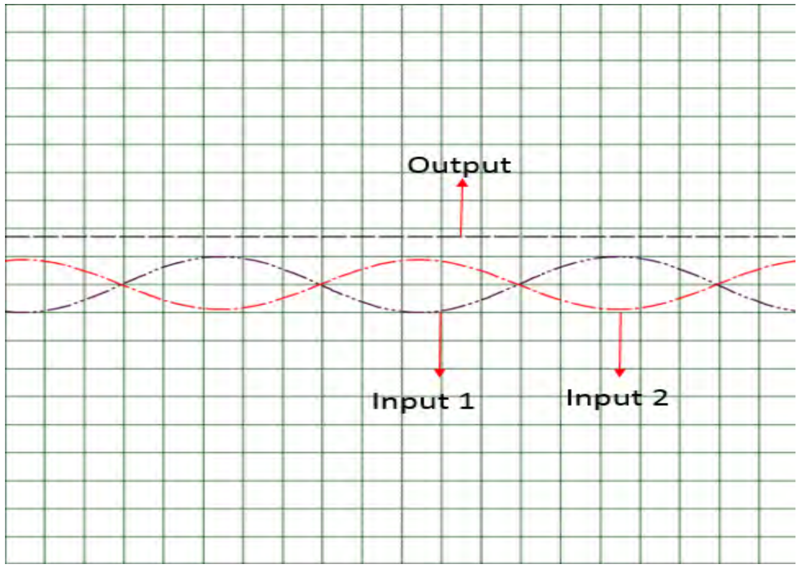


Figure 3.12: Simulation of the circuit

3.2.3 Constructed EMG Sensor Circuit

Figure 3.13 shows the constructed EMG sensor circuit which consists of preamplifier circuit, driven right leg, second order band pass filter, high precision rectifier, smoother and inverting amplifier. The constructed EMG circuit is tested before the experiment is carried by using the function generator and digital oscilloscope as shown in Figure 3.14.

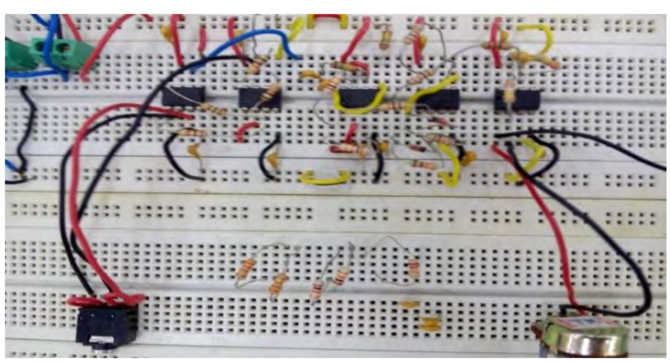


Figure 3.13: Constructed EMG sensor circuit

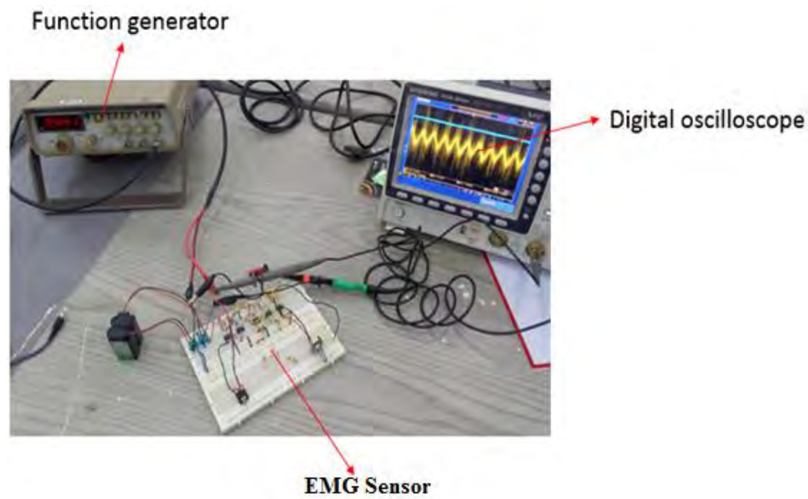


Figure 3.14: Testing the EMG sensor circuit using function generator

Figure 3.15 shows the graph from the EMG sensor circuit and function generator. The yellow line is the input from function generator to EMG sensor circuit which is a sinusoidal wave with noise. The green line is the output from EMG sensor circuit which is amplified, rectified and smoothed.

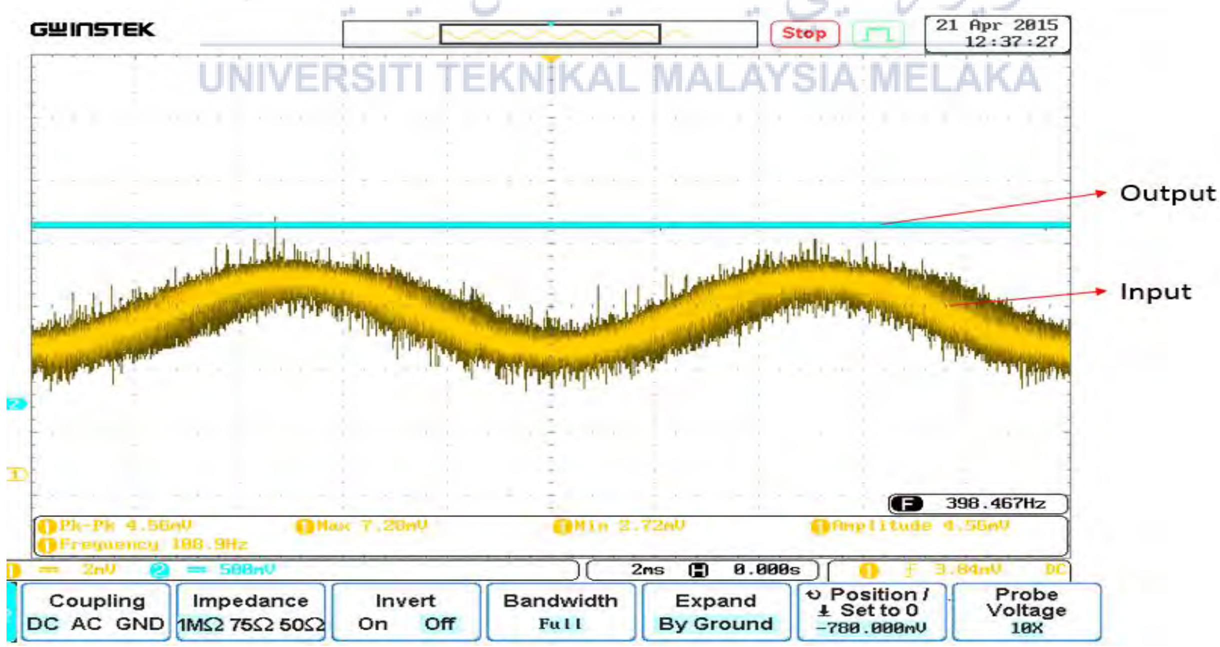


Figure 3.15: Signal Graph

3.3 Experiment Set Up for Performance Analysis

3.3.1 Experiment Set Up for Getting the EMG Signal Value Based on the Walking Motion

According to Sharif Muhammad Taslim Reza [23], the EMG signal from lower limb muscle is affected by the motion of the lower limb which means that the EMG signal is perpendicular to the motion of the lower limb. In his experiment, he recorded 5 subjects who are male with around 170 cm of height and 56 kg of weight. The subjects perform a sit and stand motion. Therefore, based on that article, a similar procedure is made.

This experiment objective is to study the behavior of EMG signal value based on the walking motion on 5 male subject which is around 166 cm of height and 56 kg of weight and find their average EMG value so that the value can be used to on and off the lower limb rehabilitation device.

Material:

1. EMG sensor
2. AgCl electrode
3. Cables
4. Digital oscilloscope

The procedures are:

1. The skin is cleaned at the rectus femoris.
2. The electrodes is placed at the muscle as shown in Figure 3.16.
3. The impedance of the muscle is measured and check based on the Table 3.4 whether it need to run a second clean or not.
4. After succeed, a simple warm up like walking is needed for at least 3 minutes.
5. Cables is connected with the electrodes.
6. The leg is Initial Contact motion as shown in Figure 3.17 and the EMG signal is recorded.
7. Walking motion for Loading Response phase is performed as shown in Figure 3.18.

8. The readings is recorded.
9. Step 1 until step 8 are repeated for 5 subjects.
10. The EMG signal value is analyzed by taking the maximum and minimum voltage value and find their average value and compared with the walking motion.

Precaution: Make sure the power cable of the digital oscilloscope is not share by other device so that there is no disturbance which could affect the accuracy of the result.

Table 3.4: Impedance range for EMG signal [9]

Impedance range (kΩ)	Recommendation
1 -5	Very good condition
5 - 10	Good and recommended if feasible
10 - 30	Acceptable for easy conditions
30 - 50	Less good, attention is needed
>50	Should be avoided or requires a second cleaning run

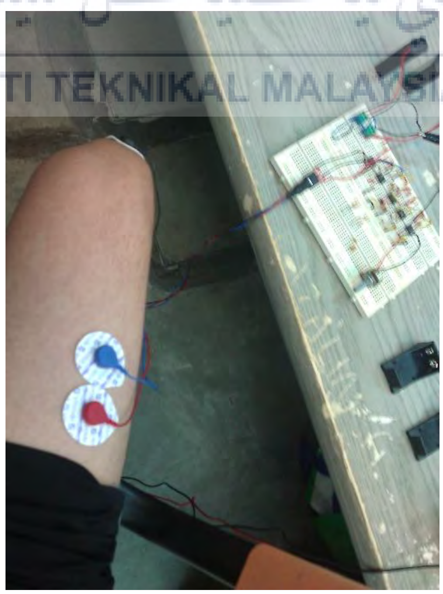


Figure 3.16: AgCl electrode is placed at rectus femoris muscle

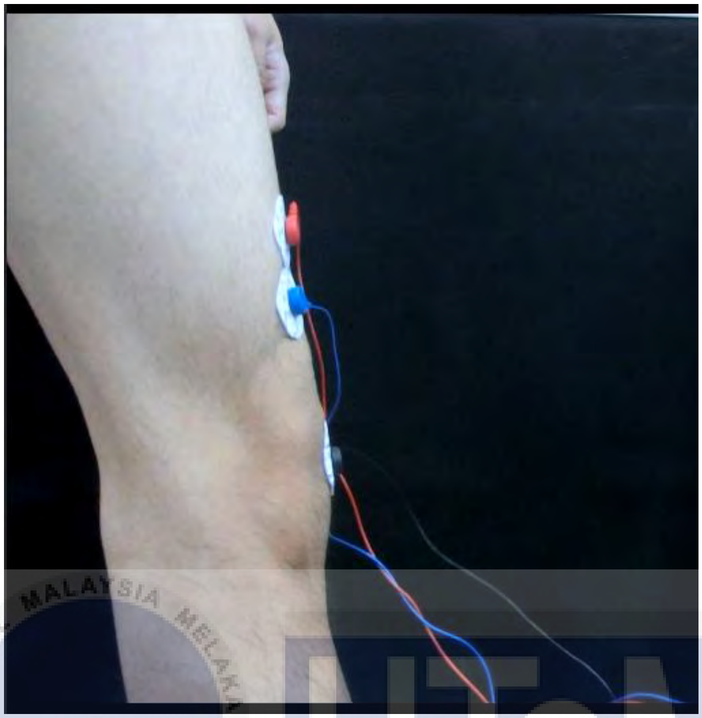


Figure 3.17: The leg is at Initial Contact phase

Figure 3.18: Walking motion at Loading Response phase

3.3.2 Experiment Set Up for Analyzing the Repeatability Performance of the Trajectory Planning of Lower Limb Rehabilitation System

According to Zhiqiang Zhang [26], the desired angle motion of the lower limb rehabilitation device can be calculated and compare it with the measured angle from the device. The measured value will have similar result with the desired angle. In his experiment, the walking motion of the device is repeated for 10 times. Therefore, based on that article, a similar procedure is made.

This experiment objective is to analyze the repeatability performance of the lower limb rehabilitation system by analyzing the trajectory of the device. The angles of the device of each joint are collected and compared with the desired value.

Material:

1. Sonxun SXSV50 servo motor with connected to the hardware of hip and knee joint
2. Arduino UNO circuit board
3. Wires
4. A piece of A4 paper
5. Protector
6. Toothpick
7. Lithium Ion battery 12V

The procedures are:

1. The reference point is drawn and set as shown in Figure 3.19 to increase the accuracy of the measurement so that every time the hardware is placed at the same place.
2. The circuit is constructed as shown in Figure 3.20.
3. The toothpick is placed at the hardware of hip joint as shown in Figure 3.21 so that it can be act as a point to measure the angle of the joint when the hardware rotate.

4. The hardware of the circuit is set up by placing the hardware on top of the reference point as shown Figure 3.21.
5. The hip joint hardware is placed at the center of reference point and the circuit is on.
6. The measurement is recorded for one complete gait cycle of the hip.
7. Step 4 until step 6 are repeated for 10 times.
8. Step 3 until step 7 are repeated by replacing the hip joint hardware with knee joint hardware on the reference point.

Precaution: Make sure the hardware of the joints in place at the exact center of the reference point by marking the point on the reference paper so that the joint place at the same time when carry out the experiment and the tooth pick is place properly by using the tape to stick to the hardware so that the tooth pick will not move every time the hardware rotate which could affect the accuracy of the result.

Figure 3.19: Reference point

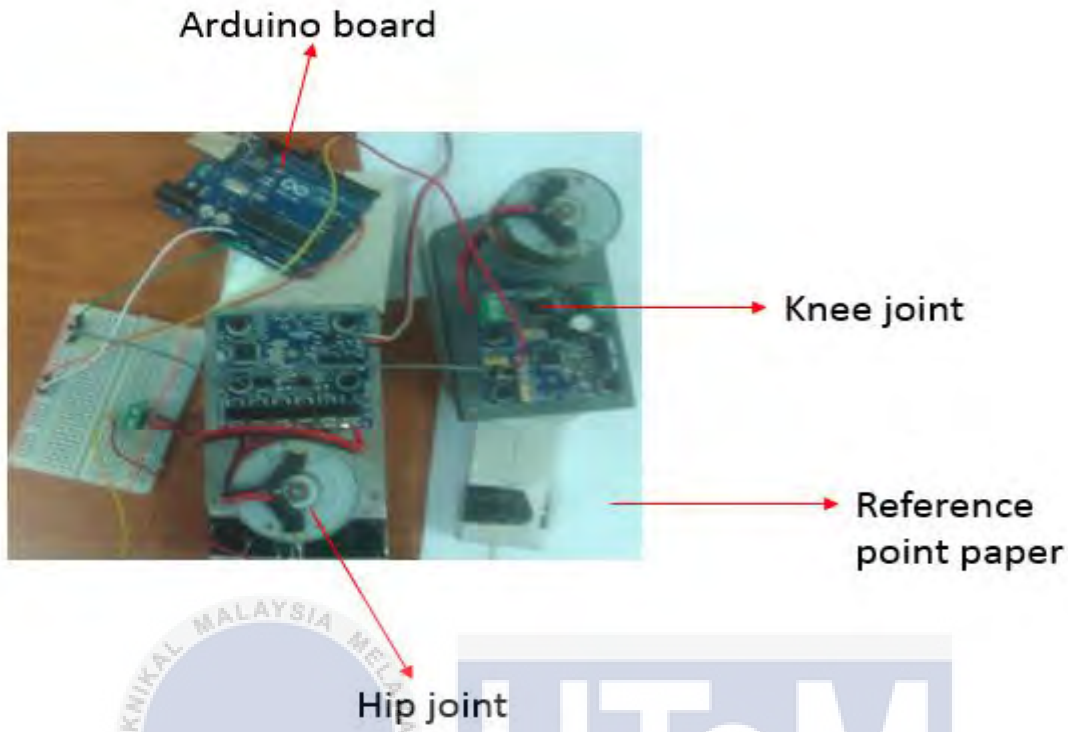


Figure 3.20: Circuit for the hardware

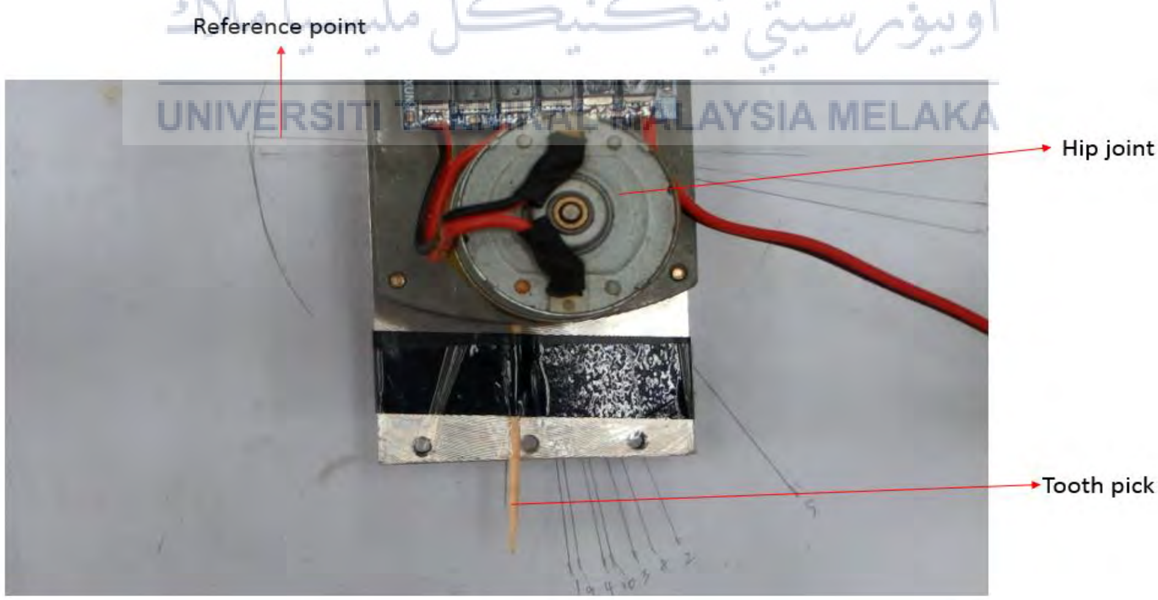


Figure 3.21: Setting of the hardware of hip joint

3.3.3 Experiment Set Up for Analyzing the Accuracy Performance of the EMG Sensor

According to Konrad, P [9], an accuracy analysis can be carried out by comparing the EMG signal from the own signal conditioning with desired EMG signal. In his experiment, a subject perform a sit and stand motion for 3 cycles. Therefore, based on that article, a similar procedure is made. The muscle chosen is rectus femoris. This experiment objective is to analyze the accuracy performance of the EMG sensor and compare the value with the muscle sensor.

Material:

1. Muscle sensor V3
2. EMG sensor
3. AgCl electrode
4. Cables
5. Digital oscilloscope

The procedures are:

1. The cleaning and warm up process is carried out exactly same as the Experiment 3.3.1 step 1 until step 8.
2. Step 1 is repeated for 3 times.
3. The muscle sensor V3 is used and step 2 and step 3 are repeated.
4. Both readings is analyzed by comparing their peak to peak voltage value of each of the circuit.

Precaution: Make sure the power cable of the digital oscilloscope is not share by other device so that there is no disturbance which could affect the accuracy of the result and make sure the electrode is placed at the same place of the muscle when using EMG sensor and Muscle sensor.

Chapter 4

RESULT AND DISCUSSION

This chapter covers all the experimental result and discussion of the experiment. This section also covers the conclusion of the experiment.

4.1 EMG Signal Value Data Collecting Based on the Walking Motion

The objective of this experiment is to study the behavior of EMG signal value based on the walking motion on 5 male subject. Table 4.1 shows the maximum and minimum voltage of experiment result getting from 5 male subject by using the EMG sensor.

The experimental subjects who are male around 56 kg and have a height of 166cm are carried out the experiment for collecting his walking motion data. The experiment is carried out with 5 subjects as shown in Table 4.1. Table 4.2 shows the average maximum voltage of the 5 subjects is 0.512V and average minimum voltage is -0.2856V.

The data of EMG signal is collecting based on the first and second walking gain phases due to the complexity of collecting the whole walking gain phases data. The average behavior of the subjects with the same characteristic EMG value based on the first and second walking motion is 0.512V for maximum and -0.2856V for minimum. The average of the maximum and minimum value can be implemented in the arduino program code which as shown in Appendices D.

Therefore, the objective of this experiment is achieved. The accuracy of the average general EMG value can be increased by increasing the number of subject with same characteristic.

Table 4.1: Experiment result of EMG sensor when perform walking motion

Subject	Maximum voltage (V)	Minimum voltage (V)
1	0.5	-0.26
2	0.56	-0.28
3	0.44	-0.348
4	0.56	-0.28
5	0.5	-0.26

Table 4.2: Average voltage value for experimental subject

Average maximum (V)	Average minimum (V)
0.512	-0.2856

4.2 Repeatability Performance of the Trajectory Planning of Lower Limb Rehabilitation System

This experiment objective is to analyze the repeatability performance of the lower limb rehabilitation system by analyzing the trajectory of the device. According to Murray MP [27], the average time for a healthy human walking for one step is 1.03 seconds. By using the equation 3.6, 3.7 and 3.8 to find the value of a0, a2 and a3 for finding the general equation of the line. Table 4.3 shows the value of a0, a2 and a3 for hip and knee joint at each phase.

After finding the value of a0, a2 and a3, by using the equation of 3.4 to plot the graph of the motion of hip and knee joint for one gait cycle which is as shown in Figure 4.1 and Figure 4.2. The results show the trend of motion of hip and knee joint which are same as the trend as shown in Figure 3.4. However, the graph in Figure 3.4 is using the degree of motion versus percentage of gait cycle but Figure 4.1 and 4.2 are using the degree of motion versus time. The calculated of the desired trajectory planning is shown in the Appendices G.

Appendices H shows the measurement of the hip joint angle and Appendices I shows the measurement of the knee joint angle which both are in radian. Both Appendices H and I contain the desired angle and average angle which calculated from the measurement of hip and knee joint angle.

The unit from result of mean square error is squared compare to root mean square error. This cause the result of root mean square error is more easily to understand. Therefore by using the root mean square error formula, average angle formula and excel function to calculate the confidence level which is used to find the consistency of the hardware:

$$avaerage = (\sum_{i=1}^{n} a_n) / 10 \tag{4.1}$$

$$RMS\ error = \sqrt{\frac{(\sum_{i=1}^{n}(a_n - desired\ value)^2)}{number\ of\ sample}} \tag{4.2}$$

Table 4.3: The value of a0, a2 and a3 for each joint at each phase

	Gait cycle (%)										
	IC (0%)	**LR (0%-10%)**	**MST (10%-30%)**	**TST (30%-50%)**		**PSW (50%-60%)**	**ISW (60%-73%)**		**MSW (73%-87%)**	**TSW (87%-100%)**	
	0%	**10%**	**30%**	**40%**	**50%**	**60%**	**66.50%**	**73%**	**87%**	**93.50%**	**100%**
time (sec)	0	0.13	0.309	0.412	0.515	0.618	0.68495	0.7519	0.8961	0.96305	1.03
time interval (sec)	0	0.13	0.179	0.103	0.103	0.103	0.06695	0.06695	0.1442	0.06695	0.06695
a0 (hip)	30		30			-10				35	
a2 (hip)	0		-809.58			929.5139				-836.6236	
a3 (hip)	0		1401.8702			-1626.0192				4165.415	
a0 (knee)	5		15		5			60			0
a2 (knee)	1775.1479		-377.2446		2214.7145			-2327.2973			3346.4943
a3 (knee)	-9103.3227		891.8312		-5409.3289			5579.2816			-33323.3194

Based on Table 4.4 and Table 4.5, the root mean square error for all phases are smaller than 1 which is relatively small. The highest value of root mean square error for the hip joint is 0.2699 radian which is at Terminal Stance 2 phase while the highest value of root mean square error for the knee joint is 0.4037 radian which is at Initial Swing 2 phase. The confidence level for hip joint of each phase is between 0.0156 radian and 0.0612 radian while the confidence level for knee joint of each phase is between 0.0161 radian and 0.0377 radian. This shows that the hip joint and knee joint have very high consistency and accuracy.

Table 4.4: Hip joint RMS error and confidence level

Hip angle (degree)	**RMS error**	**Confidence level (95%)**
Phases		
Initial Contact	0.1118	0.0612
Loading Response	0.1118	0.0612
Mid Stance	0.0761	0.0410
Terminal Stance 1	0.1942	0.0156
Terminal Stance 2	0.2699	0.0179
Pre Swing	0.2225	0.0169
Initial Swing 1	0.1786	0.0179
Initial Swing 2	0.1668	0.0179
Mid Swing	0.0691	0.0222
Terminal Swing 1	0.0603	0.0234
Terminal Swing 2	0.0774	0.0175

Table 4.5: Knee joint RMS error and confidence level

Knee angle (degree) Phases	RMS error	Confidence level (95%)
Initial Contact	0.0529	0.0377
Loading Response	0.0518	0.0262
Mid Stance	0.0515	0.0222
Terminal Stance 1	0.0317	0.0208
Terminal Stance 2	0.1525	0.0202
Pre Swing	0.3935	0.0161
Initial Swing 1	0.3050	0.0161
Initial Swing 2	0.4037	0.0169
Mid Swing	0.1235	0.0244
Terminal Swing 1	0.0830	0.0306
Terminal Swing 2	0.0611	0.0289

Figure 4.1 and Figure 4.2 shows the pattern of the graph of desired angle is same as the measured angle. However, there is some difference between the desired angle and measured angle and deviate more when the angle is at its peak value as shown in the Figure 4.1 Box A and Figure 4.2 Box B. This is due to the shaft of the servo motor does not properly fit in the joint and cause the accuracy of the servo motor reduce. This means that the hole that used to place the shaft of the servo motor is too big and cause the shaft of the servo motor slips every time it rotate.

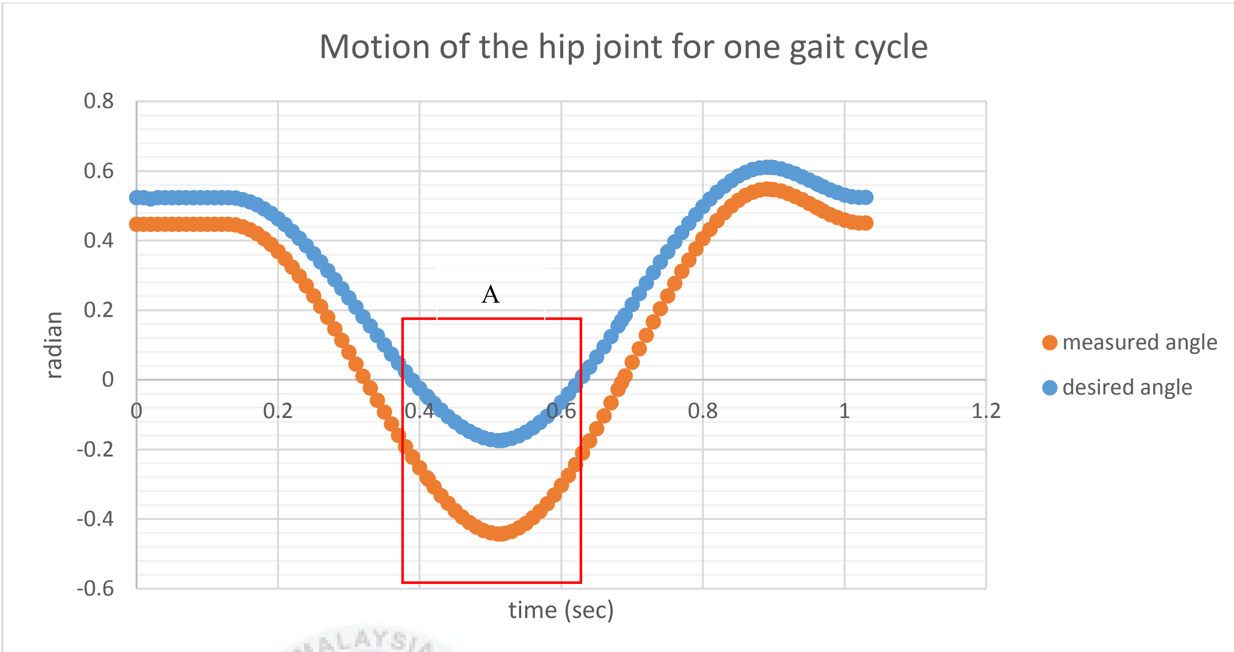


Figure 4.1: Graph of the desired and measured hip joint angle

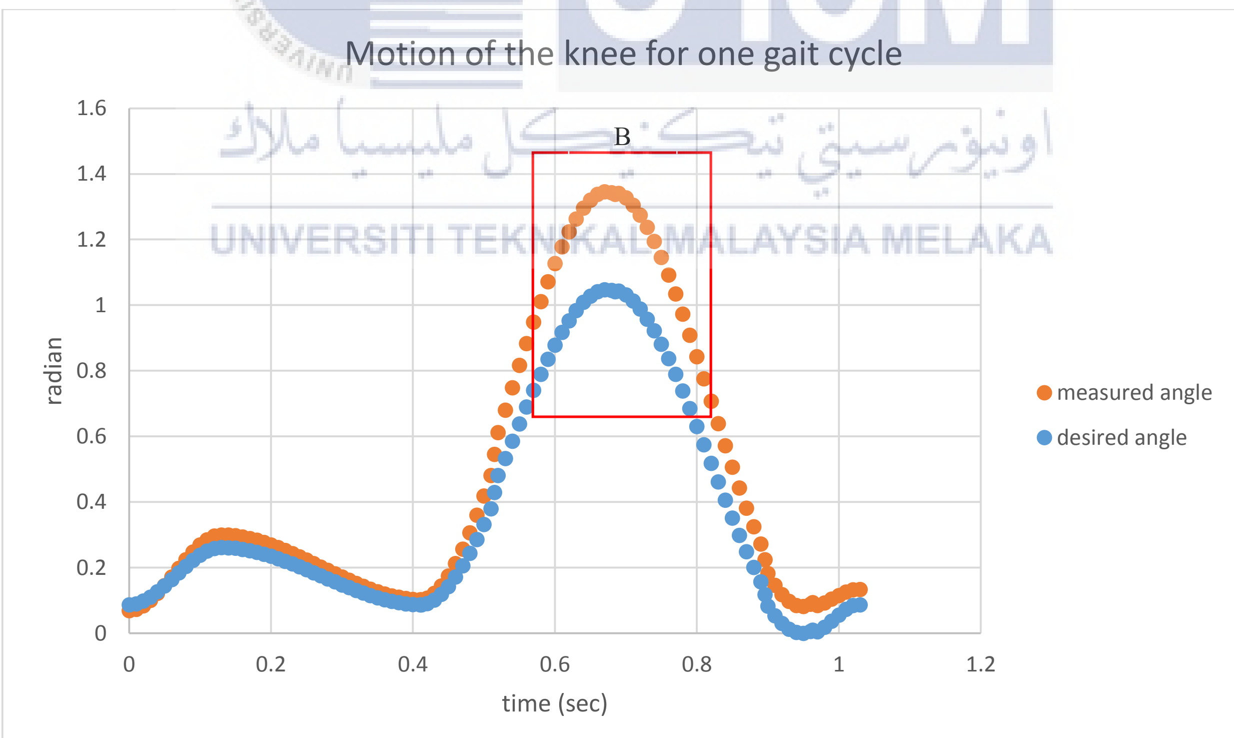


Figure 4.2: Graph of the desired and measured knee joint angle

The lower limb rehabilitation device having a relatively small error and relatively high consistency. This shows that the performance of the lower limb rehabilitation device is high. Therefore, it can be used on the post stroke patients in order to help them to regain their walking ability. As a conclusion, the objective of this experiment is achieved.

In order to increase the accuracy and reduce the error of the device, some improvement of the device need to be taken. The accuracy of the device can be improve by replacing the hardware of joint that properly fit the shaft of the servo motor.

4.3 Analyze the Performance of the EMG Sensor

This experiment objective is to analyze the accuracy performance of the EMG sensor and compare the value with the muscle sensor. Table 4.6 shows the experiment result of muscle sensor and EMG sensor by recording their peak to peak voltage.

Table 4.6: Experiment result of muscle sensor and. EMG sensor

Circuit	**Peak to peak voltage (V)**		
	1	2	3
EMG Sensor	0.76	0.76	0.78
Muscle Sensor V3	0.74	0.73	0.76

By using the accuracy equation,

$$accuracy = 100 - \frac{Theoretical\ Value - Measured\ Value}{Theoretical\ Value}\ x\ 100\% \qquad (4.3)$$

Based on the results show in Table 4.7, the average voltage of EMG sensor is 0.77V while the muscle sensor V3 is 0.74V. The muscle sensor V3 having a lower voltage than EMG sensor when perform same motion during experiment. The accuracy of the EMG sensor is around 96.86% when compare to the muscle sensor V3 as shown in Table 4.8.

Table 4.7: Average peak to peak voltage of muscle sensor and EMG sensor

Circuit	Average (V)
EMG Sensor	0.766666667
Muscle Sensor V3	0.743333333

Table 4.8: Accuracy of EMG sensor

Circuit	Accuracy (%)
EMG Sensor	96.86098655

The EMG sensor shows a very smooth graph as shown in Figure 4.3 Box C. The muscle sensor V3 shows there is a very high peak graph as shown in Figure 4.4 Box D. This shows that the EMG sensor shows a smoother graph than muscle sensor.

Figure 4.3: Example of experiment result from EMG sensor

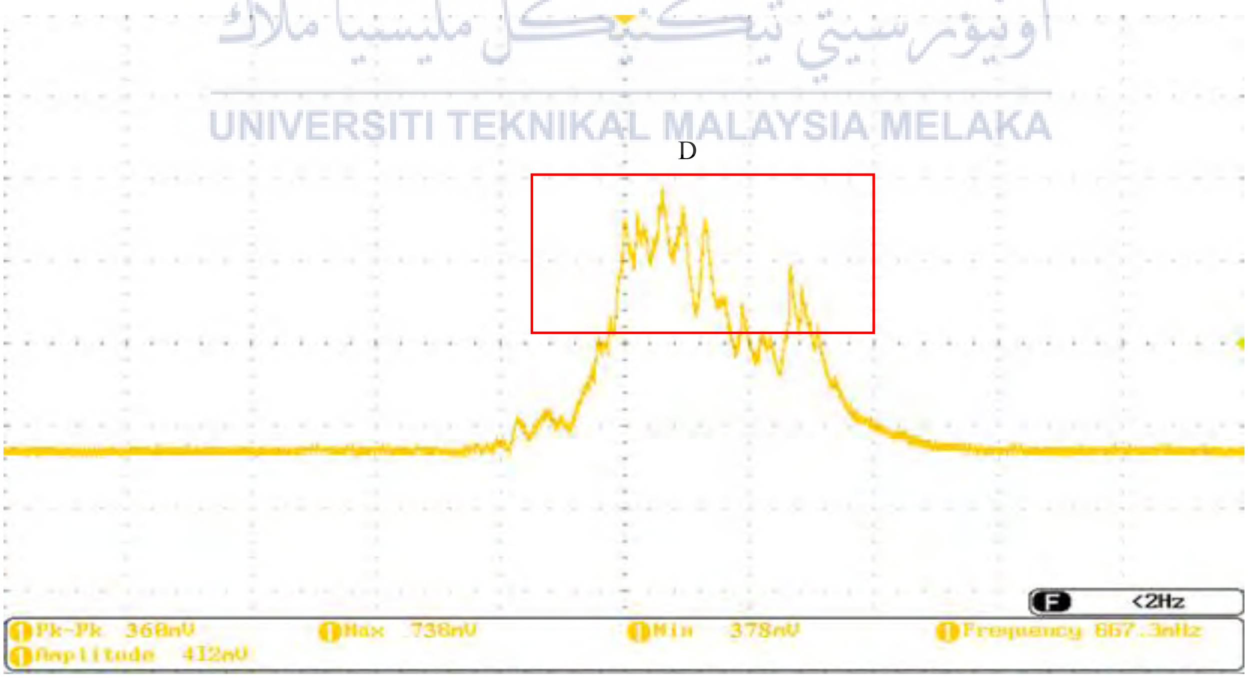


Figure 4.4: Example of experiment result from muscle sensor V3

Table 4.9 shows the characteristic of EMG sensor and muscle sensor. Both of the circuit is linear envelop type EMG circuit. However, the muscle sensor do not have bandpass filter and driven right leg. Muscle sensor is using the AD8221 as the IC at the preamplifier circuit which is purposely made for medical application and is high accuracy and very low noise disturbance while the EMG sensor using the AD620 as the IC at the preamplifier circuit which is cheap and is general purpose IC but high accuracy. This causes the EMG sensor having more noise disturbance compare to muscle sensor and this causes the EMG sensor having a higher value of peak to peak voltage compare to muscle sensor. Therefore, the accuracy of the EMG sensor is lower than the muscle sensor.

Table 4.9: Characteristic of EMG sensor and muscle sensor

Characteristic	**EMG Sensor**	**Muscle Sensor V3**
Type	Linear envelop	Linear Envelop
Preamplifier circuit	Yes	Yes
Bandpass filter	Yes	No
Driven right leg	Yes	No
Inverting amplfier	Yes	Yes
High precision rectifier	Yes	Yes
Smoother	Yes	Yes
IC used	AD620, OPA2604	AD8221, TL084

The EMG sensor is using the OPA2604 IC which is made for medical application and having a high accuracy, faster response and lower noise compare to TL084 which is general purpose IC used by the muscle sensor. This causes the EMG signal conditioning circuit having a smoother graph compare to the muscle sensor.

The accuracy of EMG sensor is relatively high when comparing to the muscle sensor V3. The reduction in the accuracy of the circuit is due to different component used in the circuit. However, the EMG sensor shows a smoother graph than muscle sensor. With these high accuracy and smoother graph, the EMG sensor can act as a Muscle sensor and the value getting from the EMG sensor can be used in the application. Therefore, the objective of this experiment is achieved. The accuracy of EMG signal conditioning can be improved by using a high precision and fast response amplifier IC.

Chapter 5

CONCLUSION AND FUTURE WORK

This chapter discuss about the conclusion based on the objectives of this project. This chapter also discuss about the recommendation that can be done in the future of the project.

5.1 Conclusion

The objectives of this project is to design a cubic polynomial and EMG based trajectory planning lower limb rehabilitation device for the post stroke patient to regain their walking ability and to analyze overall system performance in terms of accuracy and repeatability. In order to achieve these objectives, three experiments are conducted.

The first experiment is conducted to study the behavior of EMG signal value based on the walking motion on 5 male subject. From the study, we can see that the average behavior of the male subjects with the characteristic of around 166 cm of height and 56 kg of weight EMG value based on the first and second walking motion is 0.512V for maximum and -0.2856V for minimum. The average EMG value can be put inside the controller as the on off switch of the device.

The second experiment is conducted to analyze the repeatability performance of the lower limb rehabilitation device by analyzing the trajectory of the device. From the experiment, we can see that the performance of the lower limb rehabilitation device is high with a relative small root mean square error which is 0.2699 radian and 0.4037 radian and relative high consistency which is in the range of 0.0156 radian and 0.0612 radian. This shows that the performance of the lower limb rehabilitation device can be accepted and used in this project.

The third experiment is conducted to analyze the accuracy performance of the EMG sensor and compare the value with the muscle sensor. From the experiment, we can see that the performance of the EMG sensor is also high with the accuracy of the EMG sensor is around 96.86% when compare to the muscle sensor V3 relatively high accuracy compare with muscle sensor V3. This shows that the EMG sensor can be used in the lower limb rehabilitation system due to its high performance.

As a conclusion of this project, with the high performance of lower limb rehabilitation device and EMG sensor, the overall of the cubic polynomial and EMG based trajectory planning lower limb rehabilitation system performance is good and is suitable to be used on the post stroke patients in order to help them to regain their walking ability. Therefore, the lower limb rehabilitation system is successfully implemented and tested.

5.2 Future Work

This project can be further improved by including the ankle in the device. This means that the lower limb rehabilitation device will be a 3 DOF which cover hip, knee and ankle. This can help the post stroke patients who are unable to move their hip, knee and ankle. By having the actuator at the hip, knee and ankle joint, the device can help the post stroke patients to perform training at hip, knee and ankle joint. This project can help more post stroke patients in order to help them to regain their walking ability.

This project also can be improved by collecting the data for a full walking gait cycle. This can allow the device perform the walking gait by mapping the EMG signal with angle of the device.

This can allow the patients to stop at their desired angle or desired phase at any time. This can provide a more freedom to the patients rather than just follow the instruction of the device.

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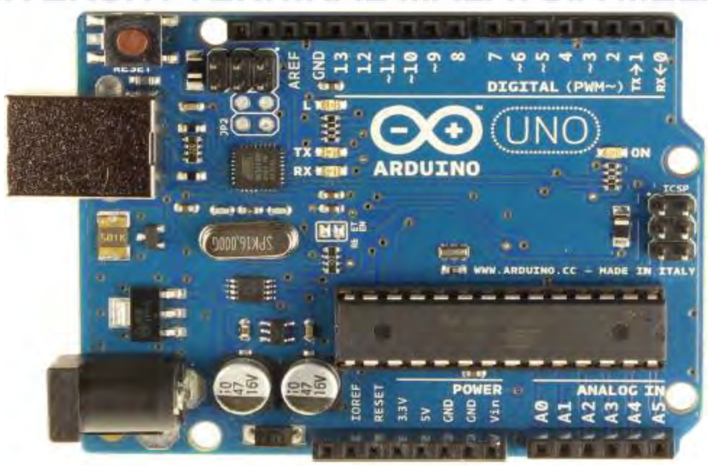
APPENDICES

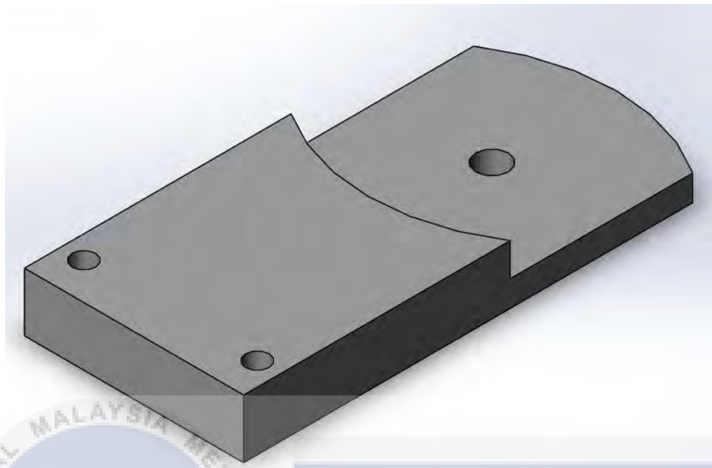
Appendices A: Component Used

Sonxun SXSV50 12 - 24 V High Power Servo Motor

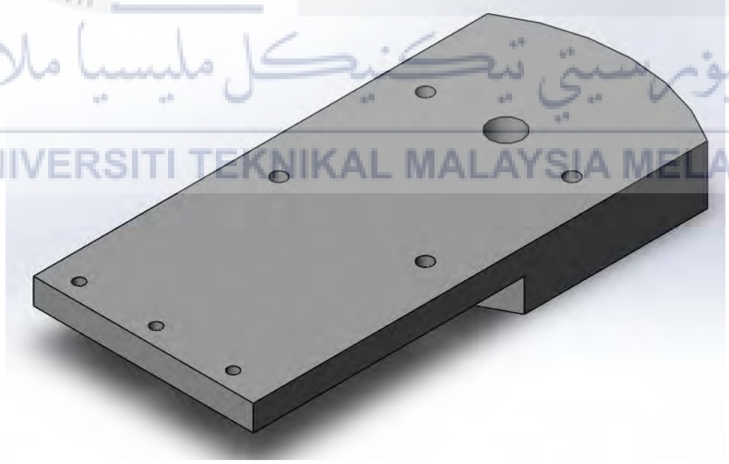


Arduino UNO board

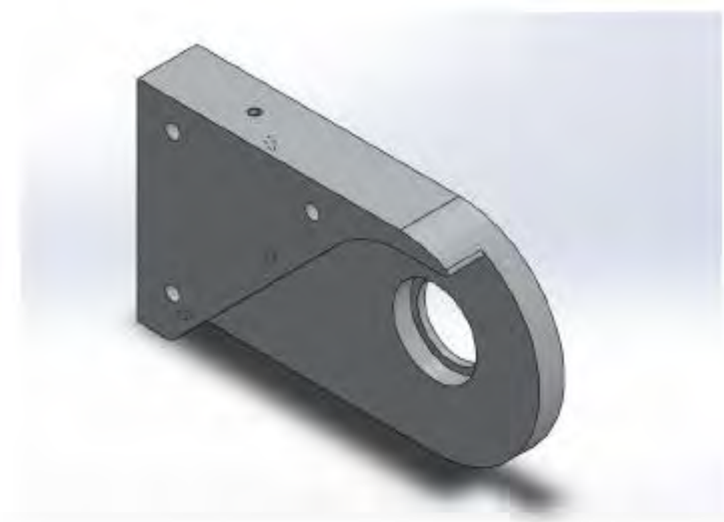
Appendices B: Lower Limb Design Drawing



Lower hip joint

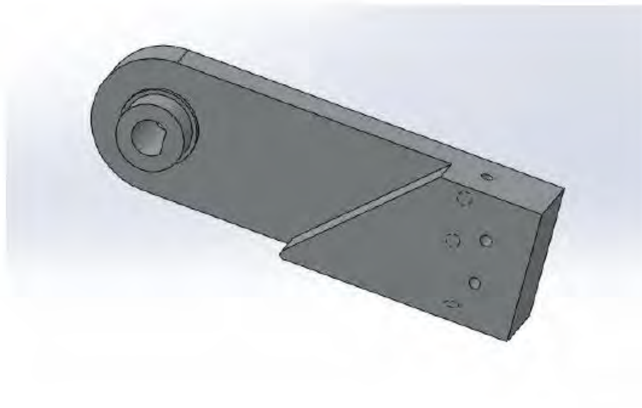


Upper hip joint



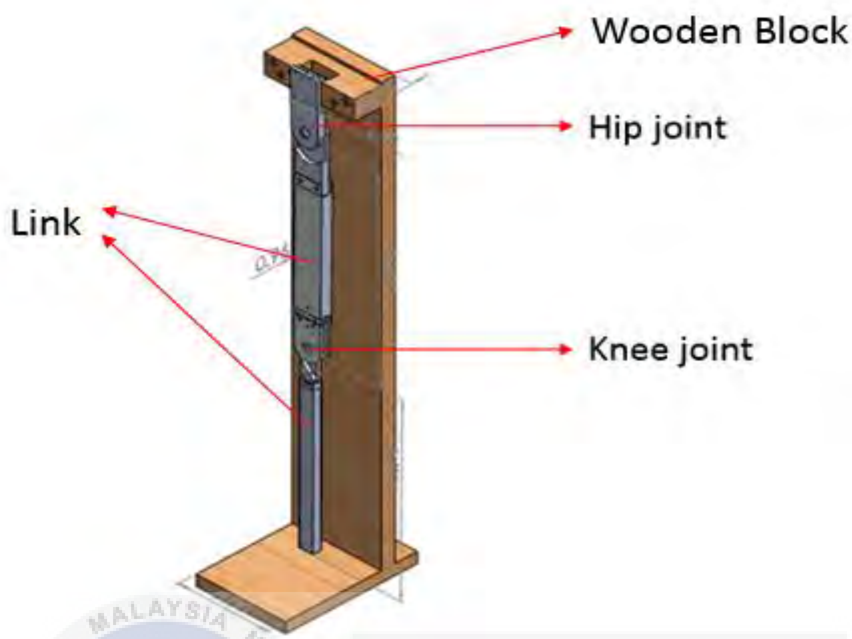
Upper knee joint

Motor plate



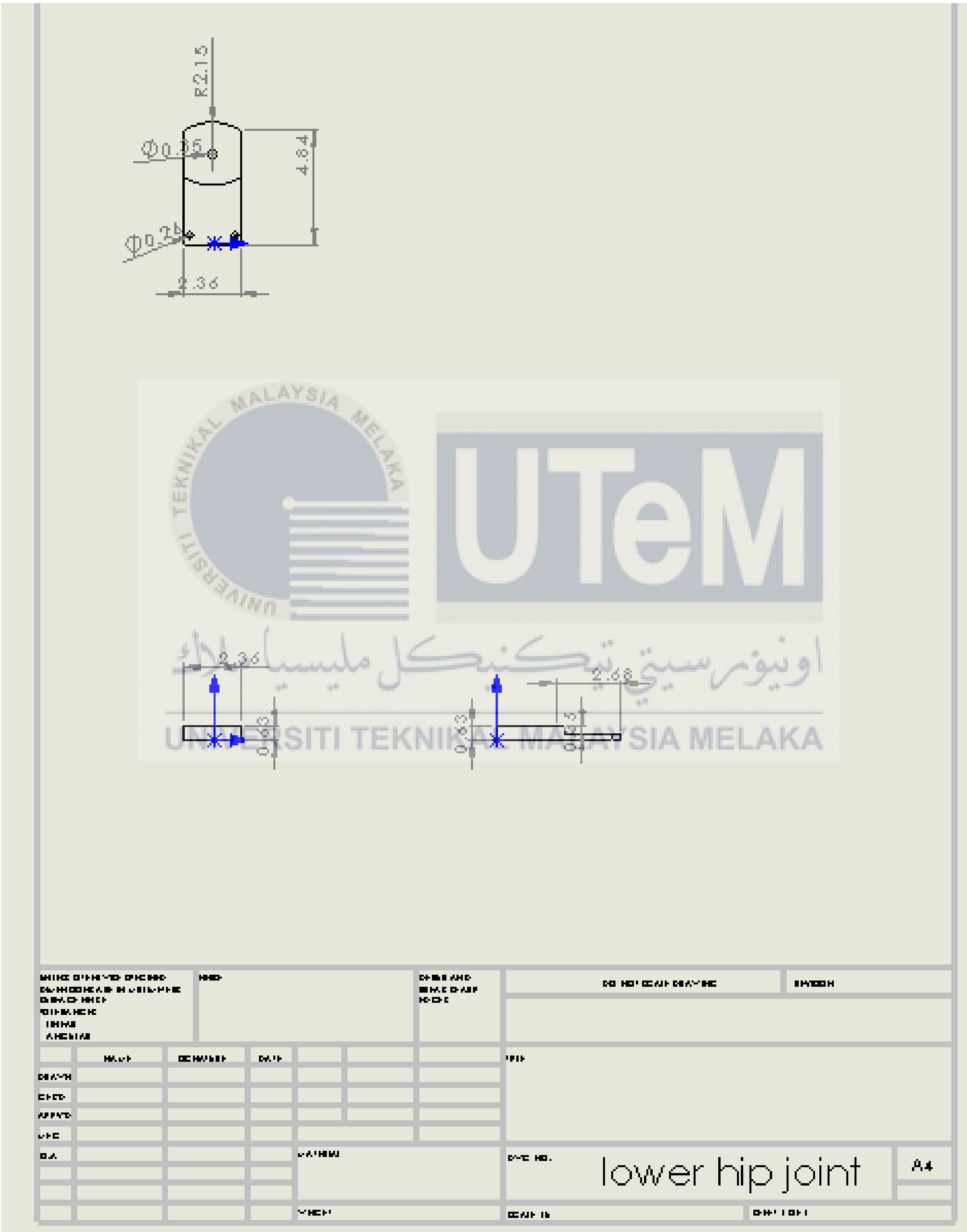
Lower knee joint

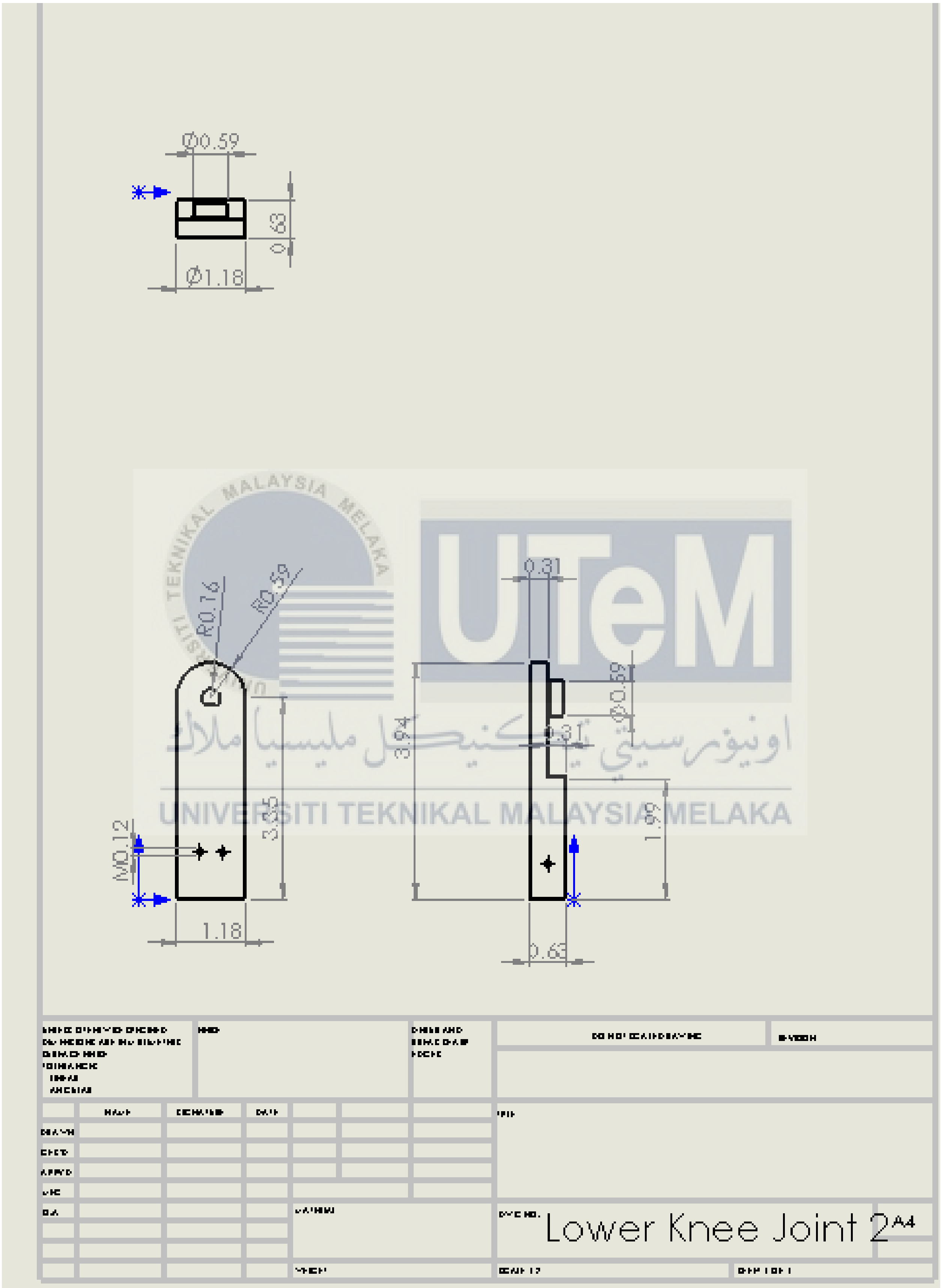
Knee joint connector



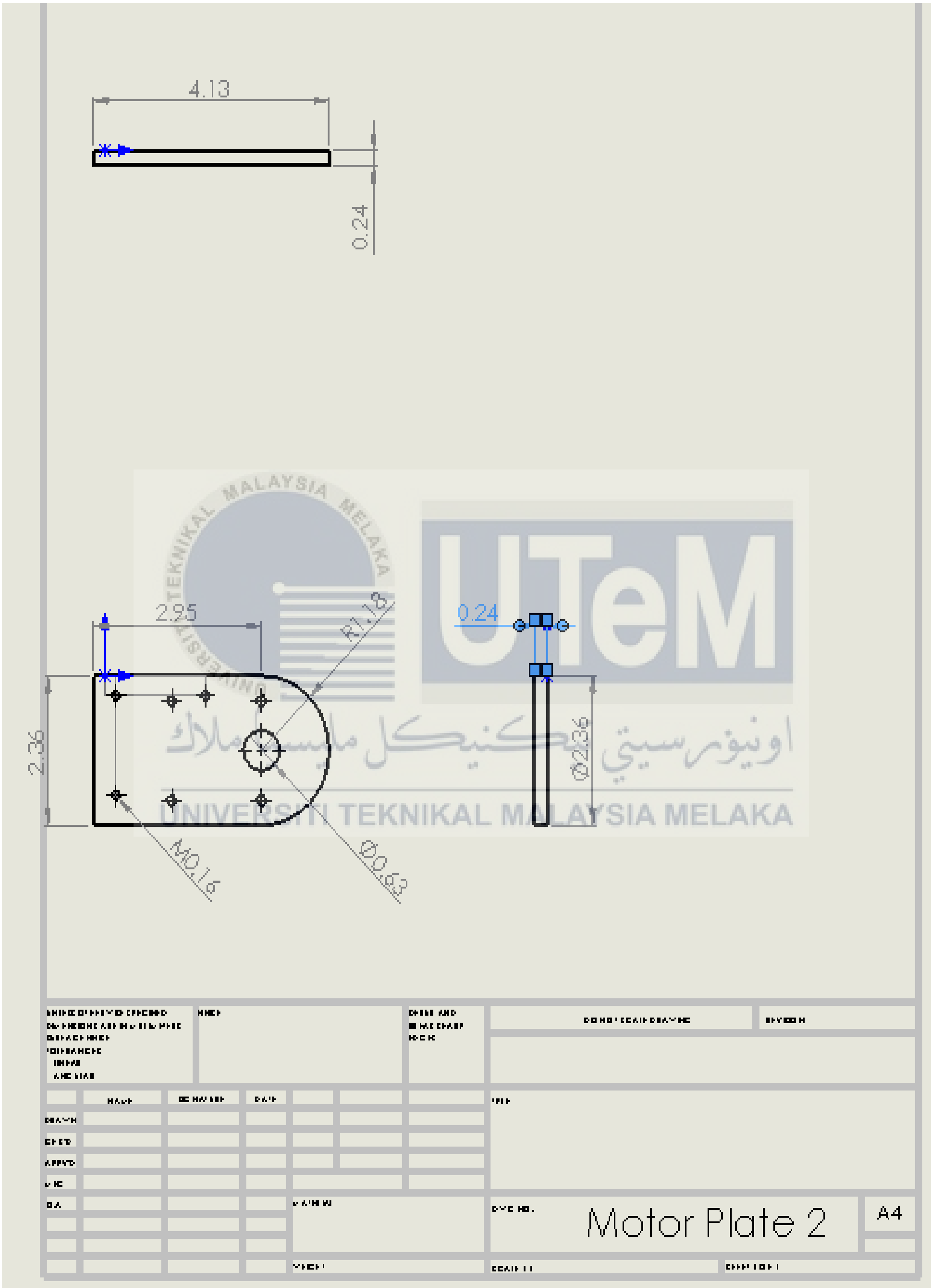
Assembly drawing of the device

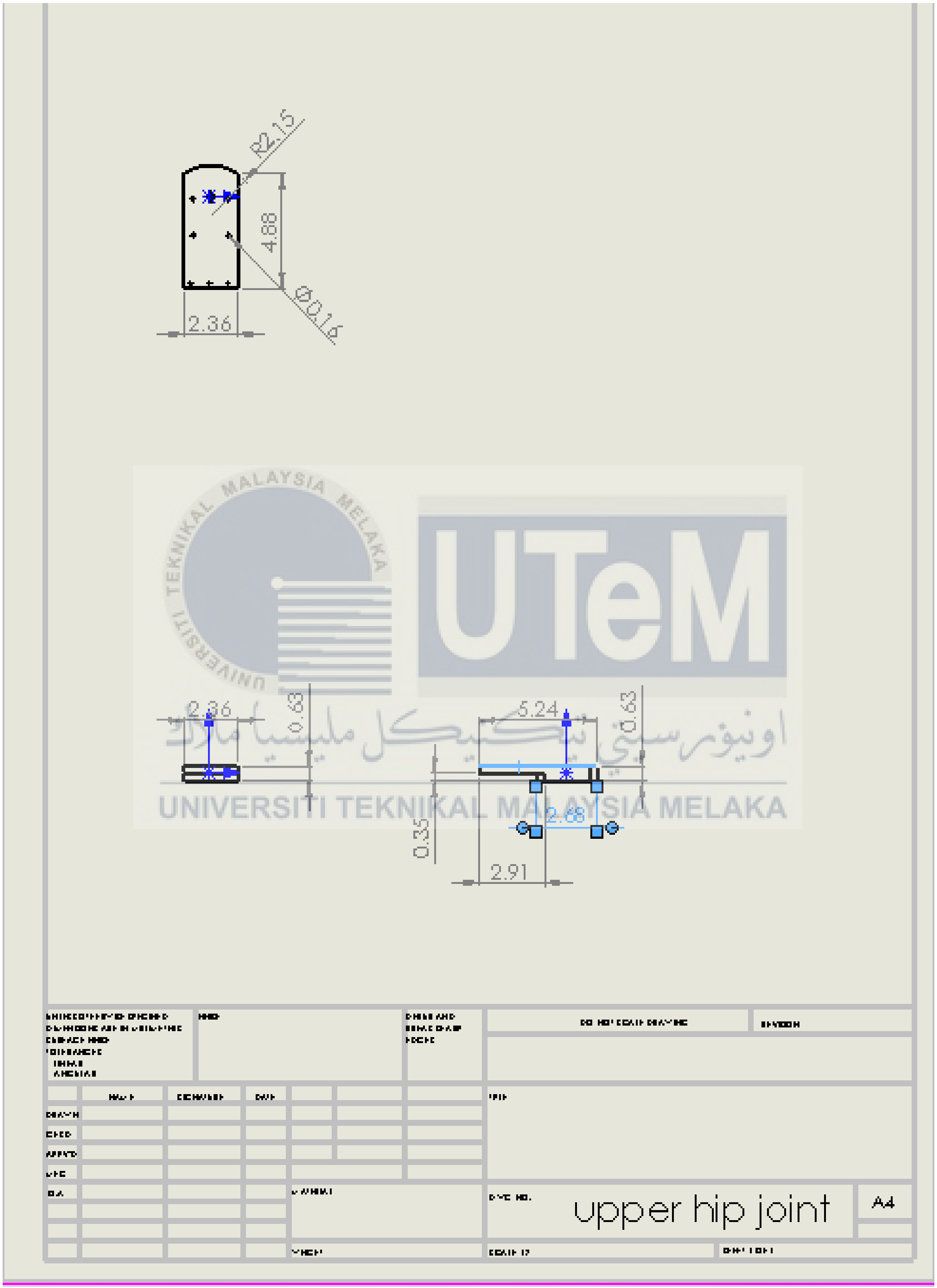
Appendices C: Dimension of Hardware Drawings

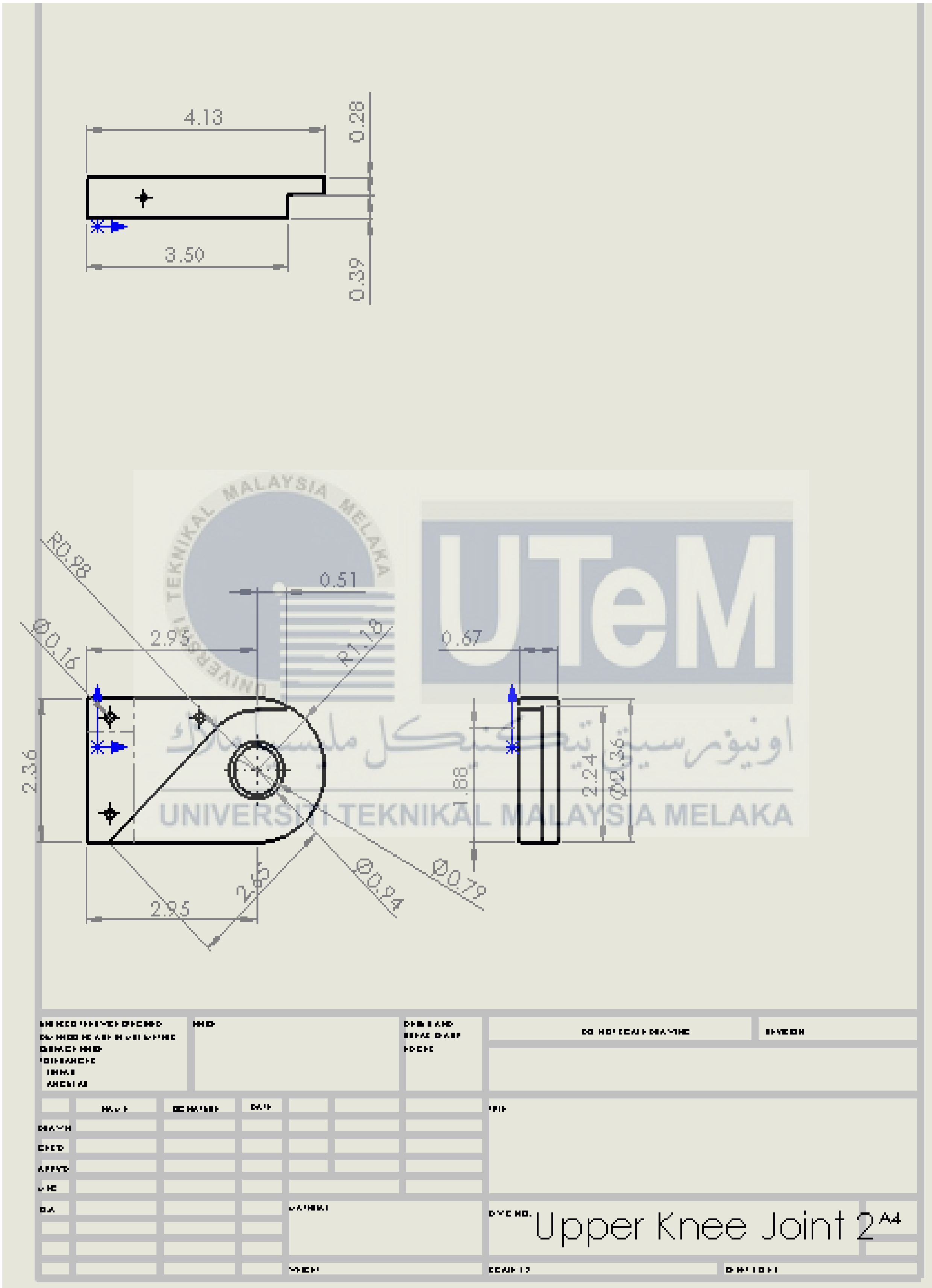




Ø0.59
0.63
Ø1.18
R0.16
R0.59
0.31
3.94
Ø0.59
0.31
3.35
1.99
M0.12
1.18
0.63
Lower Knee Joint 2
A4


4.13
0.24
2.95
R1.18
0.24
2.36
Ø2.36
M0.16
Ø0.63
Motor Plate 2
A4


R2.15
4.88
2.36
Ø0.16
2.36
0.63
5.24
0.63
2.68
0.35
2.91
DO NOT SCALE DRAWING
upper hip joint
A4


4.13
0.28
3.50
0.39
R0.98
0.51
Ø0.16
2.95
R1.18
0.67
2.36
1.88
2.24
Ø2.36
2.95
2.65
Ø0.94
Ø0.79
Upper Knee Joint 2
A4

Appendices D: Arduino Program Code

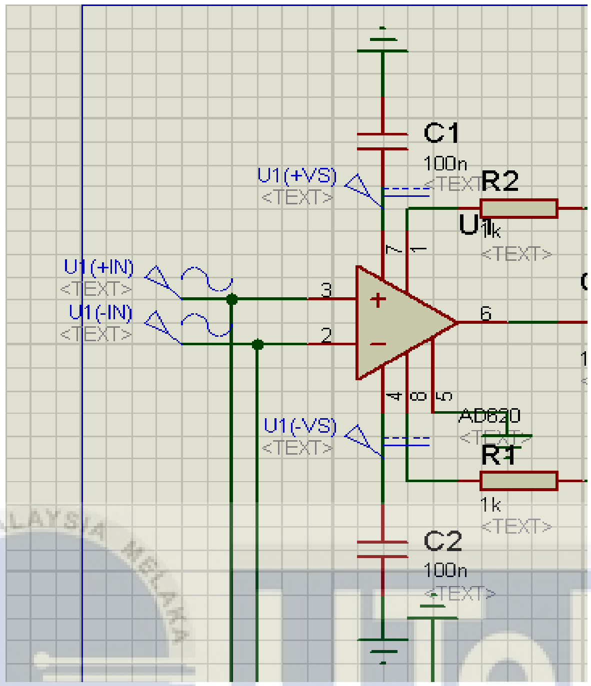
```
#include <Servo.h>
Servo hipservo;  //servo at hip joint
Servo kneeservo;  //servo at knee joint
int hipangle = 0;  //initial hip joint angle
int kneeangle = 0;  //initial knee joint angle
const int emg = 1;   //EMG signal at analog pin 0;
double sensor = 0;
double sensor 1;
int inputPin = 2; // input
double count = 0; //count time (ms)

void setup() {
  hipservo.attach(9);  //hip servo motor at pin 9
  kneeservo.attach(10);  //knee servo motor at pin 10
  pinMode (inputPin, INPUT);
  digitalWrite (inputPin, HIGH);
  Serial.begin(9600);
}

void loop() {
 sensor = analogRead(emg);
```

```
 Serial.println(sensor);
 hipservo.write(91);
 kneeservo.write(65);
sensor1 = sensor *9/1023
if (sensor1 <= 0.512 && sensor1 > 0)
  {
    hipservo.attach(9,500,2500);
    kneeservo.attach(10,500,2500);
    hipservo.write(127);  //IC
    kneeservo.write(78);  //IC
    delay (5000);
    hipangle = 30 + 97;
    kneeangle = 78;//**
    hipservo.write(hipangle);
    kneeservo.write(kneeangle);
    delay (1300);
    hipangle = 12 + 97;
    kneeangle = 78;//**
    hipservo.write(hipangle);
    kneeservo.write(kneeangle);
    delay (1790);
    hipangle = -3 + 97;
    kneeangle = 78;//**
```

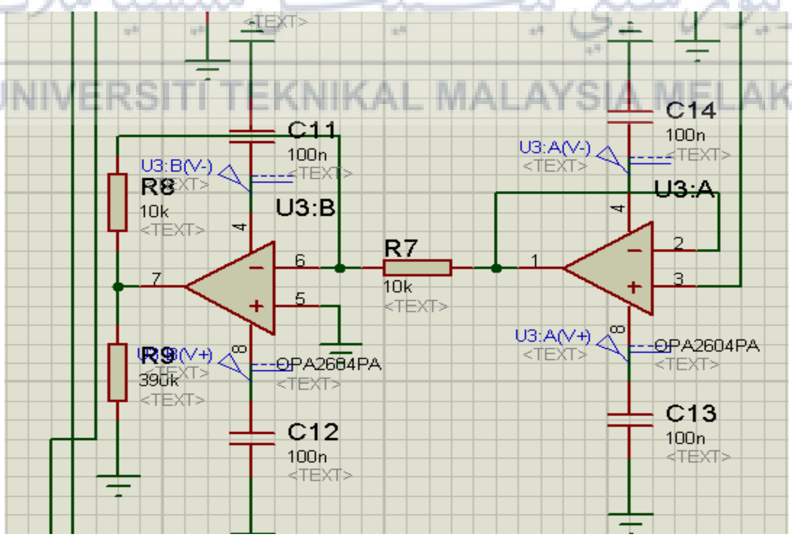
```
hipservo.write(hipangle);
kneeservo.write(kneeangle);
delay (1030);
hipangle = -10 + 97;
kneeangle = 79;//**
hipservo.write(hipangle);
kneeservo.write(kneeangle);
delay (1030);
  hipangle = -1 + 97;
  kneeangle = 80;//**
  hipservo.write(hipangle);
  kneeservo.write(kneeangle);
  delay (1030);
  hipangle = 10 + 97;
  kneeangle = 81;//**
  hipservo.write(hipangle);
  kneeservo.write(kneeangle);
  delay (670);
  hipangle = 21 + 97;
  kneeangle = 79;//**
  hipservo.write(hipangle);
  kneeservo.write(kneeangle);
  delay (670);
```

```
      hipangle = 35 + 97;
      kneeangle = 78;//**
      hipservo.write(hipangle);
      kneeservo.write(kneeangle);
      delay (1442);
      hipangle = 32 + 97;
      kneeangle = 77;//**
      hipservo.write(hipangle);
      kneeservo.write(kneeangle);
      delay (670);
      hipangle = 30 + 97;
      kneeangle = 78;//**
      hipservo.write(hipangle);
      kneeservo.write(kneeangle);
      delay (670);
   }
   else
   {
     hipservo.attach(9,0,0);
     kneeservo.attach(10,0,0);
   }
 }
```

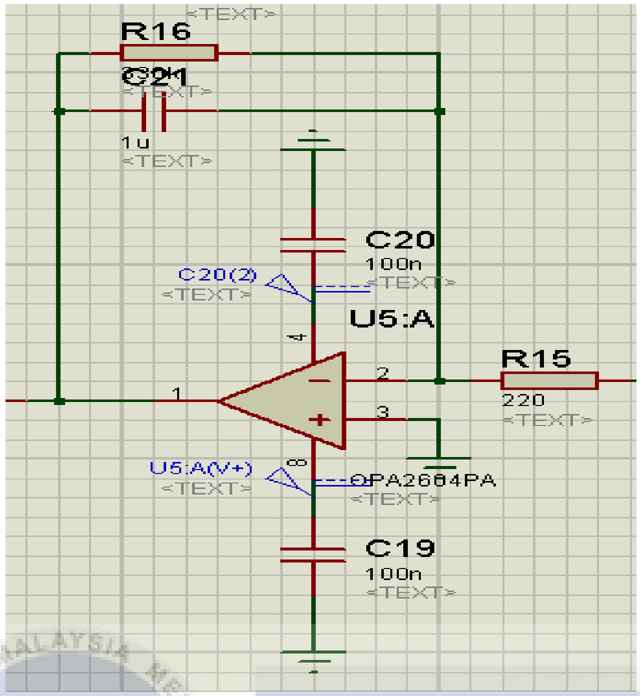
Appendices E: Types of circuit of EMG sensor



Preamplifier circuit

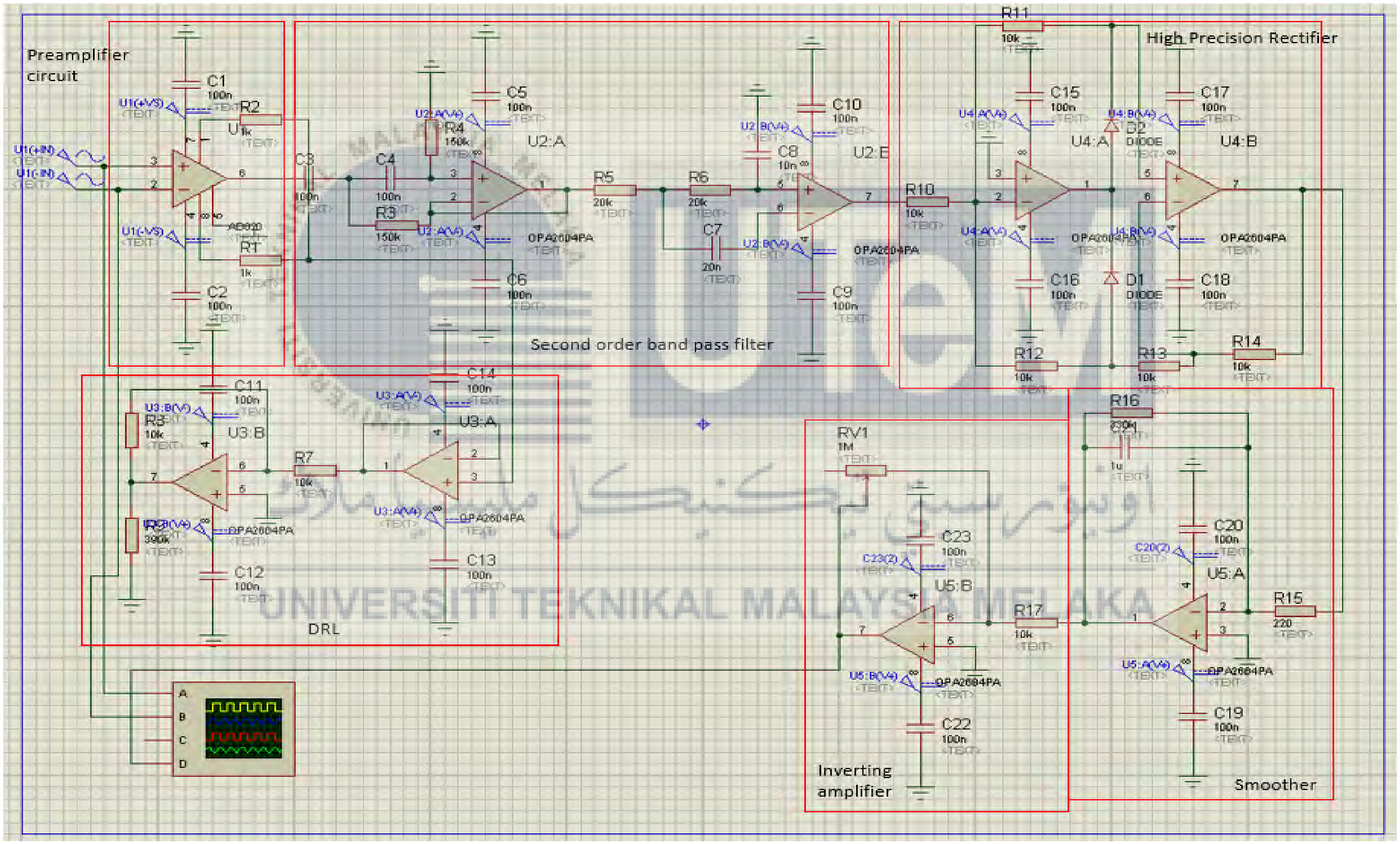


Driven right leg circuit



Smoother

Appendices F: Schematic Diagram



Appendices G: Calculated Value for Trajectory Planning

Time (knee)	Time (hip)	Actual time	a0	a1	a2	angle	a0	a1	a2	angle
		0	30	0	0	30	5	1775.1479	-9103.3227	5
		0.01	30	0	0	30	5	1775.1479	-9103.3227	5.168411467
		0.02	30	0	0	30	5	1775.1479	-9103.3227	5.637232578
		0.03	30	0	0	30	5	1775.1479	-9103.3227	6.351843397
		0.04	30	0	0	30	5	1775.1479	-9103.3227	7.257623987
		0.05	30	0	0	30	5	1775.1479	-9103.3227	8.299954413
		0.06	30	0	0	30	5	1775.1479	-9103.3227	9.424214737
		0.07	30	0	0	30	5	1775.1479	-9103.3227	10.57578502
		0.08	30	0	0	30	5	1775.1479	-9103.3227	11.70004534
		0.09	30	0	0	30	5	1775.1479	-9103.3227	12.74237574
		0.1	30	0	0	30	5	1775.1479	-9103.3227	13.6481563
		0.11	30	0	0	30	5	1775.1479	-9103.3227	14.36276708
		0.12	30	0	0	30	5	1775.1479	-9103.3227	14.83158813
	0	0.13	30	-809.58	1401.8702	30	5	1775.1479	-9103.3227	14.99999954
0.01	0.01	0.14	30	-809.58	1401.8702	29.92044387	15	-377.2446	891.8312	14.96316737
0.02	0.02	0.15	30	-809.58	1401.8702	29.68738296	15	-377.2446	891.8312	14.85623681
0.03	0.03	0.16	30	-809.58	1401.8702	29.3092285	15	-377.2446	891.8312	14.6845593
0.04	0.04	0.17	30	-809.58	1401.8702	28.79439169	15	-377.2446	891.8312	14.45348584
0.05	0.05	0.18	30	-809.58	1401.8702	28.15128378	15	-377.2446	891.8312	14.1683674
0.06	0.06	0.19	30	-809.58	1401.8702	27.38831596	15	-377.2446	891.8312	13.83455498
0.07	0.07	0.2	30	-809.58	1401.8702	26.51389948	15	-377.2446	891.8312	13.45739956
0.08	0.08	0.21	30	-809.58	1401.8702	25.53644554	15	-377.2446	891.8312	13.04225213

0.09	0.09	0.22	30	-809.58	1401.8702	24.46436538	15	-377.2446	891.8312	12.59446368
0.1	0.1	0.23	30	-809.58	1401.8702	23.3060702	15	-377.2446	891.8312	12.1193852
0.11	0.11	0.24	30	-809.58	1401.8702	22.06997124	15	-377.2446	891.8312	11.62236767
0.12	0.12	0.25	30	-809.58	1401.8702	20.76447971	15	-377.2446	891.8312	11.10876207
0.13	0.13	0.26	30	-809.58	1401.8702	19.39800683	15	-377.2446	891.8312	10.58391941
0.14	0.14	0.27	30	-809.58	1401.8702	17.97896383	15	-377.2446	891.8312	10.05319065
0.15	0.15	0.28	30	-809.58	1401.8702	16.51576193	15	-377.2446	891.8312	9.5219268
0.16	0.16	0.29	30	-809.58	1401.8702	15.01681234	15	-377.2446	891.8312	8.995478835
0.17	0.17	0.3	30	-809.58	1401.8702	13.49052629	15	-377.2446	891.8312	8.479197746
0.18	0.18	0.31	30	-809.58	1401.8702	11.94531501	15	-377.2446	891.8312	7.978434518
0.19	0.19	0.32	30	-809.58	1401.8702	10.3895897	15	-377.2446	891.8312	7.498540141
0.2	0.2	0.33	30	-809.58	1401.8702	8.8317616	15	-377.2446	891.8312	7.0448656
0.21	0.21	0.34	30	-809.58	1401.8702	7.280241922	15	-377.2446	891.8312	6.622761883
0.22	0.22	0.35	30	-809.58	1401.8702	5.74344189	15	-377.2446	891.8312	6.237579978
0.23	0.23	0.36	30	-809.58	1401.8702	4.229772723	15	-377.2446	891.8312	5.89467087
0.24	0.24	0.37	30	-809.58	1401.8702	2.747645645	15	-377.2446	891.8312	5.599385549
0.25	0.25	0.38	30	-809.58	1401.8702	1.305471875	15	-377.2446	891.8312	5.357075
0.26	0.26	0.39	30	-809.58	1401.8702	-0.088337365	15	-377.2446	891.8312	5.173090211
0.27	0.27	0.4	30	-809.58	1401.8702	-1.425370853	15	-377.2446	891.8312	5.05278217
0.28	0.28	0.41	30	-809.58	1401.8702	-2.69721737	15	-377.2446	891.8312	5.001501862
0.282	0.282	0.412	30	-809.58	1401.8702	-2.943024049	15	-377.2446	891.8312	5.000000016
0.01	0.29	0.42	30	-809.58	1401.8702	-3.895465692	5	2214.7145	-5409.3289	5.216062121
0.02	0.3	0.43	30	-809.58	1401.8702	-5.0117046	5	2214.7145	-5409.3289	5.842611169
0.03	0.31	0.44	30	-809.58	1401.8702	-6.037522872	5	2214.7145	-5409.3289	6.84719117

0.04	0.32	0.45	30	-809.58	1401.8702	-6.964509286	5	2214.7145	-5409.3289	8.19734615
0.05	0.33	0.46	30	-809.58	1401.8702	-7.784252623	5	2214.7145	-5409.3289	9.860620138
0.06	0.34	0.47	30	-809.58	1401.8702	-8.488341659	5	2214.7145	-5409.3289	11.80455716
0.07	0.35	0.48	30	-809.58	1401.8702	-9.068365175	5	2214.7145	-5409.3289	13.99670124
0.08	0.36	0.49	30	-809.58	1401.8702	-9.515911949	5	2214.7145	-5409.3289	16.4045964
0.09	0.37	0.5	30	-809.58	1401.8702	-9.822570759	5	2214.7145	-5409.3289	18.99578668
0.1	0.38	0.51	30	-809.58	1401.8702	-9.979930386	5	2214.7145	-5409.3289	21.7378161
0.11	0.385	0.515	30	-809.58	1401.8702	-9.999994498	5	2214.7145	-5409.3289	24.59822868
0.12	0.01	0.52	-10	929.5139	-1626.0192	-9.908674629	5	2214.7145	-5409.3289	27.54456846
0.13	0.02	0.53	-10	929.5139	-1626.0192	-9.641202594	5	2214.7145	-5409.3289	30.54437946
0.14	0.03	0.54	-10	929.5139	-1626.0192	-9.207340008	5	2214.7145	-5409.3289	33.5652057
0.15	0.04	0.55	-10	929.5139	-1626.0192	-8.616842989	5	2214.7145	-5409.3289	36.57459121
0.16	0.05	0.56	-10	929.5139	-1626.0192	-7.87946765	5	2214.7145	-5409.3289	39.54008003
0.17	0.06	0.57	-10	929.5139	-1626.0192	-7.004970107	5	2214.7145	-5409.3289	42.42921616
0.18	0.07	0.58	-10	929.5139	-1626.0192	-6.003106476	5	2214.7145	-5409.3289	45.20954366

0.19	0.08	0.59	-10	929.5139	-1626.0192	-4.88363287	5	2214.7145	-5409.3289	47.84860652
0.2	0.09	0.6	-10	929.5139	-1626.0192	-3.656305407	5	2214.7145	-5409.3289	50.3139488
0.21	0.1	0.61	-10	929.5139	-1626.0192	-2.3308802	5	2214.7145	-5409.3289	52.57311451
0.22	0.11	0.62	-10	929.5139	-1626.0192	-0.917113365	5	2214.7145	-5409.3289	54.59364767
0.23	0.12	0.63	-10	929.5139	-1626.0192	0.575238982	5	2214.7145	-5409.3289	56.34309232
0.24	0.13	0.64	-10	929.5139	-1626.0192	2.136420728	5	2214.7145	-5409.3289	57.78899249
0.25	0.14	0.65	-10	929.5139	-1626.0192	3.756675755	5	2214.7145	-5409.3289	58.89889219
0.26	0.15	0.66	-10	929.5139	-1626.0192	5.42624795	5	2214.7145	-5409.3289	59.64033545
0.27	0.16	0.67	-10	929.5139	-1626.0192	7.135381197	5	2214.7145	-5409.3289	59.98086631
0.28	0.17	0.68	-10	929.5139	-1626.0192	8.87431938	5	2214.7145	-5409.3289	59.88802879
0.28495	0.17495	0.68495	-10	929.5139	-1626.0192	9.74311976	5	2214.7145	-5409.3289	59.67173491
0.01	0.18	0.69	-10	929.5139	-1626.0192	10.63330639	60	-2327.2973	5579.2816	59.77284955
0.02	0.19	0.7	-10	929.5139	-1626.0192	12.4025861	60	-2327.2973	5579.2816	59.11371533
0.03	0.2	0.71	-10	929.5139	-1626.0192	14.1724024	60	-2327.2973	5579.2816	58.05607303
0.04	0.21	0.72	-10	929.5139	-1626.0192	15.93299918	60	-2327.2973	5579.2816	56.63339834

0.05	0.22	0.73	-10	929.5139	-1626.0192	17.67462032	60	-2327.2973	5579.2816	54.87916695
0.06	0.23	0.74	-10	929.5139	-1626.0192	19.3875097	60	-2327.2973	5579.2816	52.82685455
0.07	0.24	0.75	-10	929.5139	-1626.0192	21.06191122	60	-2327.2973	5579.2816	50.50993682
0.08	0.25	0.76	-10	929.5139	-1626.0192	22.68806875	60	-2327.2973	5579.2816	47.96188946
0.09	0.26	0.77	-10	929.5139	-1626.0192	24.25622618	60	-2327.2973	5579.2816	45.21618816
0.1	0.27	0.78	-10	929.5139	-1626.0192	25.7566274	60	-2327.2973	5579.2816	42.3063086
0.11	0.28	0.79	-10	929.5139	-1626.0192	27.17951628	60	-2327.2973	5579.2816	39.26572648
0.12	0.29	0.8	-10	929.5139	-1626.0192	28.51513672	60	-2327.2973	5579.2816	36.12791748
0.13	0.3	0.81	-10	929.5139	-1626.0192	29.7537326	60	-2327.2973	5579.2816	32.92635731
0.14	0.31	0.82	-10	929.5139	-1626.0192	30.8855478	60	-2327.2973	5579.2816	29.69452163
0.15	0.32	0.83	-10	929.5139	-1626.0192	31.90082621	60	-2327.2973	5579.2816	26.46588615
0.16	0.33	0.84	-10	929.5139	-1626.0192	32.78981172	60	-2327.2973	5579.2816	23.27392655
0.17	0.34	0.85	-10	929.5139	-1626.0192	33.5427482	60	-2327.2973	5579.2816	20.15211853
0.18	0.35	0.86	-10	929.5139	-1626.0192	34.14987955	60	-2327.2973	5579.2816	17.13393777
0.19	0.36	0.87	-10	929.5139	-1626.0192	34.60144964	60	-2327.2973	5579.2816	14.25285996

0.2	0.37	0.88	-10	929.5139	-1626.0192	34.88770237	60	-2327.2973	5579.2816	11.5423608
0.21	0.38	0.89	-10	929.5139	-1626.0192	34.99888162	60	-2327.2973	5579.2816	9.035915968
0.22	0.3861	0.8961	-10	929.5139	-1626.0192	34.97656318	60	-2327.2973	5579.2816	6.767001157
0.23	0.01	0.9	35	-836.6236	4165.415	34.92050306	60	-2327.2973	5579.2816	4.769092057
0.24	0.02	0.91	35	-836.6236	4165.415	34.69867388	60	-2327.2973	5579.2816	3.075664358
0.25	0.03	0.92	35	-836.6236	4165.415	34.35950497	60	-2327.2973	5579.2816	1.72019375
0.26	0.04	0.93	35	-836.6236	4165.415	33.9279888	60	-2327.2973	5579.2816	0.736155922
0.27	0.05	0.94	35	-836.6236	4165.415	33.42911788	60	-2327.2973	5579.2816	0.157026563
0.28	0.06	0.95	35	-836.6236	4165.415	32.88788468	60	-2327.2973	5579.2816	0.016281363
0.29	0.07	0.96	35	-836.6236	4165.415	32.32928171	60	-2327.2973	5579.2816	0.347396012
0.29305	0.07305	0.96305	35	-836.6236	4165.415	32.15927254	60	-2327.2973	5579.2816	0.547410687
0.01	0.08	0.97	35	-836.6236	4165.415	31.77830144	0	3346.4943	-33323.3194	0.301326111
0.02	0.09	0.98	35	-836.6236	4165.415	31.25993638	0	3346.4943	-33323.3194	1.072011165
0.03	0.1	0.99	35	-836.6236	4165.415	30.799179	0	3346.4943	-33323.3194	2.112115246
0.04	0.11	1	35	-836.6236	4165.415	30.42102181	0	3346.4943	-33323.3194	3.221698438

0.05	0.12	1.01	35	-836.6236	4165.415	30.15045728	0	3346.4943	-33323.3194	4.200820825
0.06	0.13	1.02	35	-836.6236	4165.415	30.01247792	0	3346.4943	-33323.3194	4.84954249
0.07	0.14	1.03	35	-836.6236	4165.415	30.0320762	0	3346.4943	-33323.3194	4.967923516

Appendices H: Measurement of the Hip Joint Angle

Hip angle (radian)	Number of times										Average angle	Desired angle
Phases	**1**	**2**	**3**	**4**	**5**	**6**	**7**	**8**	**9**	**10**		
Initial Contact	0.3666	0.3491	0.3840	0.4364	0.3666	0.4538	0.4538	0.6109	0.5062	0.5411	0.4469	0.5237
Loading Response	0.3666	0.3491	0.3840	0.4364	0.3666	0.4538	0.4538	0.6109	0.5062	0.5411	0.4469	0.5237
Mid Stance	0.0873	0.0873	0.1396	0.1571	0.0698	0.1920	0.1746	0.2095	0.2095	0.2269	0.1554	0.2086
Terminal Stance 1	-0.2095	-0.2618	-0.2444	-0.2618	-0.2618	-0.2444	-0.2793	-0.2269	-0.2269	-0.2269	-0.2444	-0.0513
Terminal Stance 2	-0.4015	-0.4713	-0.4538	-0.4364	-0.4713	-0.4538	-0.4713	-0.4189	-0.4364	-0.4189	-0.4434	-0.1746
Pre Swing	-0.2095	-0.2444	-0.2444	-0.2618	-0.2618	-0.2269	-0.2444	-0.1920	-0.2269	-0.2618	-0.2374	-0.0161
Initial Swing 1	0.0000	-0.0349	0.0000	0.0349	-0.0524	0.0000	0.0000	0.0175	-0.0175	-0.0175	-0.0070	0.1700
Initial Swing 2	0.2095	0.1920	0.2095	0.2444	0.1746	0.1571	0.1920	0.2095	0.2095	0.2269	0.2025	0.3676
Mid Swing	0.5935	0.5411	0.5237	0.5237	0.5062	0.5411	0.5237	0.5760	0.5586	0.5935	0.5481	0.6106

Terminal Swing 1	0.5411	0.5062	0.4888	0.4888	0.4713	0.5062	0.4713	0.5237	0.5237	0.5760	0.5097	0.5614
Terminal Swing 2	0.4713	0.4364	0.4364	0.4364	0.4015	0.4713	0.4364	0.4713	0.4713	0.4713	0.4504	0.5242

Appendices I: Measurement of the Knee Joint Angle

Knee angle (radian)	Number of times										Average angle	Desired angle
Phases	**1**	**2**	**3**	**4**	**5**	**6**	**7**	**8**	**9**	**10**		
Initial Contact	0.1396	0.0524	0.0698	0.0524	0.0873	0.1746	0.0175	0.0175	0.0698	0.0175	0.0698	0.0873
Loading Response	0.3317	0.3142	0.2618	0.3666	0.2793	0.3317	0.3142	0.2618	0.2793	0.2618	0.3002	0.2618
Mid Stance	0.2269	0.1746	0.1746	0.1571	0.1746	0.2444	0.1920	0.1571	0.1571	0.1571	0.1815	0.1393
Terminal Stance 1	0.1396	0.1047	0.1047	0.0698	0.1222	0.1571	0.0873	0.0698	0.0873	0.0873	0.1030	0.0873
Terminal Stance 2	0.6109	0.5935	0.6109	0.5935	0.5760	0.6109	0.5586	0.5586	0.5411	0.5411	0.5795	0.4294
Pre Swing	1.3615	1.3790	1.3092	1.3441	1.3266	1.3790	1.3441	1.3441	1.3266	1.3441	1.3458	0.9529
Initial Swing 1	1.3615	1.3790	1.3092	1.3441	1.3266	1.3790	1.3441	1.3441	1.3266	1.3441	1.3458	1.0416
Initial Swing 2	1.2917	1.3092	1.2393	1.2743	1.2743	1.3266	1.2917	1.2917	1.2743	1.2743	1.2847	0.8817
Mid Swing	0.2618	0.2444	0.2095	0.1920	0.2444	0.3142	0.2444	0.2269	0.2269	0.2095	0.2374	0.1182

Terminal Swing 1	0.0873	0.1222	0.0524	0.0175	0.0873	0.1746	0.0873	0.0698	0.0698	0.0524	0.0820	0.0096
Terminal Swing 2	0.1571	0.1571	0.1047	0.0873	0.1396	0.2269	0.1396	0.1222	0.1047	0.1047	0.1344	0.0868

Appendices J: Gantt Chart