WEARABLE POWER ASSISTED PNEUMATIC-BASED ANKLE FOOT ORTHOSIS (AFO)

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A project report submitted in partial fulfilment of the requirements for the award of Bachelor of Engineering (Hons.) Biomedical Engineering

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DECLARATION

I hereby declare that this project report is based on my original work except for citations and quotations which have been duly acknowledged. I also declare that it has not been previously and concurrently submitted for any other degree or award at UTAR or other institutions.

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APPROVAL FOR SUBMISSION

I certify that this project report entitled **"WEARABLE POWER ASSISTED PNEUMATIC-BASED ANKLE FOOT ORTHOSIS (AFO)"** was prepared by **WONG ZHEN YANG** has met the required standard for submission in partial fulfilment of the requirements for the award of Bachelor of Engineering (Hons.) Biomedical Engineering at Universiti Tunku Abdul Rahman.

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WEARABLE POWER ASSISTED PNEUMATIC-BASED ANKLE FOOT ORTHOSIS (AFO)

ABSTRACT

Design and development of wearable power assisted pneumatic-based ankle foot orthosis (AFO) is discussed in this thesis. This project is a part of a modular AFO system to develop a wearable power assisted pneumatic-based knee-ankle-foot orthosis (KAFO). AFO is used to support the ankle, hold the foot and ankle in the correct position and correct foot drop. The objective of the project is to create a low cost power assisted AFO to assist patient with lower extremities problem in rehabilitation and daily life. McKibben pneumatic artificial muscle (PAM) is chosen as the pneumatic actuator in this project. The wearable power assisted pneumatic based AFO in this project is built for left lower limb. It consists of five main parts: 1) AFO, 2) frames, 3) PAMs, 4) actuation system, and 5) control system. Fabrication of AFO, frames and McKibben PAMs are reported in this thesis. Actuation system and control system is discussed for both knee and ankle joint together as they are closely related in this project. The McKibben PAMs fabricated is tested and shown sufficient and matching properties to be use in the prototype with some limitations. On the other hand, the prototype is tested with a subject and show similar gait pattern with actual human gait pattern. However, a few problems are found during the experiments. The problems including imbalance shown by the subject during the experiment and air flow produce by a compact compressor is insufficient. A few improvements have been suggested for the prototype, including fabrication of another power assisted KAFO for right leg with actuation at hip joint, and feedback system to detect ground contact to control the duration of stance phase.

LIST OF PUBLICATIONS

NO.	TITLE
1.	Teng, C., Wong, Z., Teh, W., & Chong, Y. Z. (2012) Design and Development of Inexpensive Pneumatically-Powered Assisted Knee-Ankle-Foot Orthosis for Gait Rehabilitation-Preliminary Finding, *Engineering and Science*. doi: 10.1109/ICoBE.2012.6178949
2.	Wong, Z., Teng, C., Chong Y. (2012) Power Asissted Pneumatic-Based Knee-Ankle-Foot Orthosis for Rehabilitation, *2012 IEEE EMBS Conference on Biomedical Engineering (IECBES 2012)*, 17-18 Dec. 2012.

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LIST OF SYMBOLS / ABBREVIATIONS

AFO	ankle-foot orthosis
BSPT	British Standard Pipe thread
BWS	body weight system
CD	controlled dorsiflexion
CP	controlled plantarflexion
DOF	degree of freedom
EMG	electromyography
FF	foot flat
FO	foot off
KAFO	knee-ankle-foot orthosis
PAM	Pelvis Assist Manipulator
PAM	pneumatic artificial muscle
POGO	Pneumatically Operated Gait Orthosis
PP	powered plantarflexion
VGRF	vertical ground reaction force

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CHAPTER 1

INTRODUCTION

1.1 Background

In recent years, research on rehabilitation system using pneumatic muscle actuator has increased and primarily focused on hand and arm rehabilitation (Li, Xia, and Guan, 2008, Blaya & Herr, 2009). Hence, a project on wearable power assisted pneumatic-based device for lower leg has been proposed. This project is divided into two parts, knee orthosis and ankle-foot orthosis, but this report focused on ankle joint actuation, and hence ankle-foot orthosis is discussed.

Ankle-foot orthosis (AFO) is a brace that worn on the lower leg and foot to support the ankle, hold the foot and ankle in the correct position and correct foot drop. It is also known as foot-drop brace. Foot drop is gait pathology where a motor deficiency caused by total or partial central paralysis of the muscles innervated by the common peroneal nerve, or the anterior tibial muscle and the peroneal group. The two major complications of drop foot are slapping of the foot after heel strike (foot slap) and dragging of the toe during swing (toe drag). At heel strike, the foot generally falls uncontrolled to the ground, producing a distinctive slapping noise (foot slap). During midswing, toe drag prevents proper limb advancement and increases the risk of tripping (Blaya & Herr, 2004).

However, conventional AFO does not permit free ankle motion during stance phase of gait (Chin et al, 2009). According to Li et al. (2008), joint rehabilitation is important after osteoarthritis or operation to completely recover the functions of the joints. Hence, it is hard to achieve the ideal rehabilitation effects with the usage of traditional AFO. Ideal foot-drop AFO should permit free ankle plantarflexion (with mild resistance) and dorsiflexion during stance phase and block plantarflexion during swing. Also, traditional rehabilitation system is driven by motor, where the joints are driven inflexibly, lack of smoothness and the resistance of active movement of the joints is not yielding.

1.2 Aims and Objectives

The aims and objectives of this project are:

- To develop a low cost wearable power-assisted pneumatic-based orthotic to aid patient with lower extremities problem
- To assist patient with lower extremities problem in rehabilitation
- To study human gait cycle with the power assisted pneumatic-based arthosis

CHAPTER 2

LITERATURE REVIEW

2.1 Pneumatic Artificial Muscle

2.1.1 Introduction

Pneumatic actuators have been considered as substitute of conventional motors due to its high ratio power/weight and power/volume ratios. Pneumatic artificial muscles (PAMs) are a type of pneumatic actuators made mainly of a flexible and inflatable membrane. PAMs are contractible device operated by pressurized air. When pressurized, PAMs will inflate, shorten and generate a contraction force. PAM has many advantages. Its elasticity is useful for the biped locomotion's natural frequency. Its high power/volume ratios make gear needless to increase power contrary to motor. Also, the way PAM operates is very similar to animal's real muscle, which is very effective to implement humanoid. PAM itself does not make any noise, and hence applicable to hospital applications. Also, compared to motor, PAM is immune to water so it is suitable for underwater applications (Choi, Jin & Lee, 2006). The disadvantages of PAM included it must have antagonist structure (Choi et al, 2006). PAM can only contract so if one side of pair is contracted it has no way to return unless the other one contracts to stretch it. The most significant problem of the PAM however, is the nonlinearity. Robot will suffer loss at precision and low speed response due to PAM's elasticity and small vibration (Cho, Jin & Lee, 2008).

However, in some robot application like human robot interaction (HRI) application that need high elasticity without considering high precision in millimetre, PAM is best fitted (Choi et al, 2008). An example of HRI application is a wearable power assist robot, which the robot developed is equipped in the body to support the bodily movement by assisting and enhancing the muscular force (Noritsugu, Takaiwa, Sasaki, 2008). The robot can be used to support patients' daily life, social participation, nursing work, rehabilitation and etc (Choi et al, 2008). A novel legs rehabilitation exercise system driven by pneumatic muscle actuator is described in a work by Li et al. (2008). It showed a smooth movement of joints, and the resistance of active movement is yielding. Also, neither inertial impingement nor electromagnetic lag is among the special features of the pneumatic muscle actuators.

2.1.2 Classification

There are four general types of PAMs: (1) braided muscles, (2) pleated PAM, (3) netted muscles, and (4) embedded muscles (Daerden & Lefeber, 2001b). For the purpose of this project, only two PAMs are focused, the embedded muscles and pleated PAMs based on the work of Daerden and Lefeber (2001a).

2.1.2.1 Braided Muscles

Braided muscles are composed of a gas-tight elastic tube or bladder surrounded by a braided sleeving. The braid fibres run helically about the muscle's long axis at an angle. When pressurized the tube presses laterally against the sleeve. Thereby the internal pressure is balanced by braid fibre tension due to fibre curvature about the tube. Fibre tension is integrated at the braid's end points to balance an external load. As the pressing contact between tube and sleeving is absolutely necessary to convey load, braided muscles cannot operate at underpressure: passing through the meshes of the braid, surrounding gas would only act on the tube that, consequently, would be squeezed without transferring a noticeable amount of work to the clamps.

The general behaviour of braided muscles with regard to shape, contraction and tension when inflated will depend on the geometry of the inner elastic part and of the braid at rest (meaning neither pressurized nor loaded), and on the materials used. Usually, braided muscles have a cylindrical shape because of a cylindrical bladder and a constant pitch angle throughout the braid. Two basic types of braided muscles can be distinguished: one that has both its inner tube and braid connected to fittings at both ends and another that only has the braid connected to end fittings and whose inner tube is an unattached bladder. The former type is generally referred to as the McKibben Muscle. The latter type is referred to as the Sleeved Bladder Muscle.

McKibben Muscle as shown in Figure 2.1 is a PAM which is most frequently used and published about at the present. It is a cylindrical braided muscle that has both its tube and its sleeving connected at both ends to fittings that not only transfer fibre tension but also serve as gas closure. It was introduced as an orthotic actuator in the late 1950's: due to the similarity in length-load curves between this artificial muscle and skeletal muscle. The advantages of McKibben muscles are its simple design, ease of assembly and low cost. However, life expectancy of this muscle seems not very high. Users complain about early braid fibre failure at the point of clamping. A major disadvantage of the McKibben Muscle is its inherent dry friction and threshold pressure, and hence position control is difficult to achieve. Other disadvantages are the limited displacement and the energy needed to deform the rubber membrane, lowering the output force.

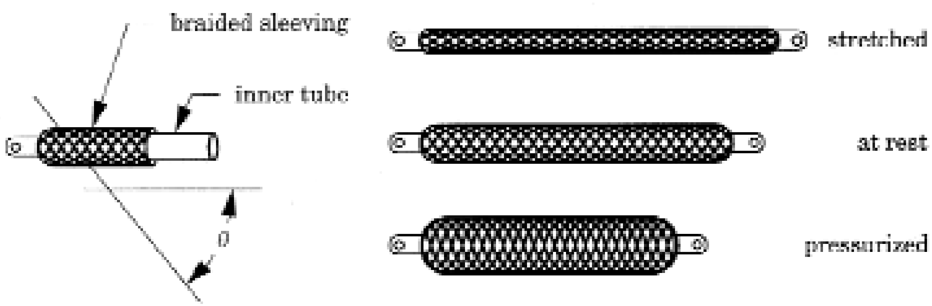


Figure 2.1: McKibben Muscle. (Daerden & Lefeber, 2001b)

Sleeved bladder muscle on the other hand differs from the McKibben type in the design of the inner tube. The inner tube does not connect to the sleeving, and hence no passive spring force is added to the muscle tension.

2.1.2.2 Pleated PAM

The pleated PAM was developed by Daerden in 1999. The pleated PAM is an artificial muscle of membrane rearranging kind. Hence, no material strain is involved during the inflation. The muscle membrane has a number of pleats in the axial direction as shown in Figure 2.2. The pleated PAM expands by unfolding the pleats. No friction is involved in the process and stresses in the parallel direction are kept negligibly small and decreased with the increasing of folds. As a result, no energy is needed to expand the membrane and the absence of friction shows practically no hysteresis.

According to Daerden (1999), material of membrane for pleated PAM should have following properties: 1) good tear resistant, 2) non brittle, 3) showing non-catastrophic failure, and 4) low creep. These properties will ensure longer life span of the pleated PAM for numerous folding and unfolding.

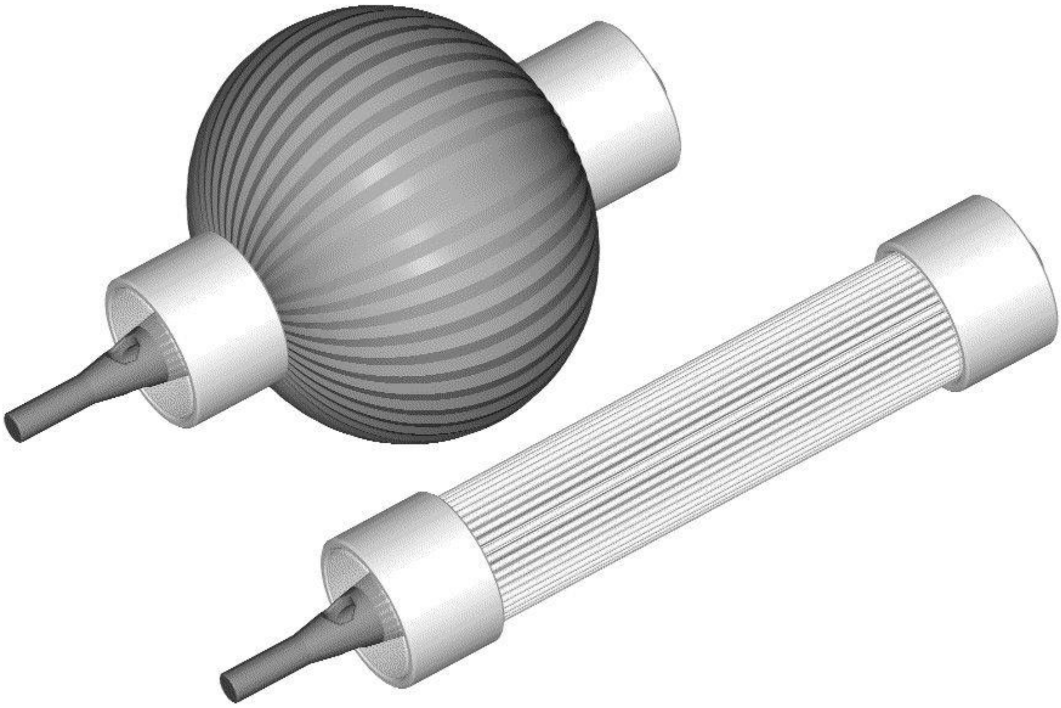


Figure 2.2: Pleated PAM. (Daerden, 1999)

2.1.3 Comparisons between McKibben Muscle and Pleated PAM

Comparisons have been made between McKibben muscle and pleated PAM by Damme et al (2008). McKibben muscle is relatively easy to fabricate, but has some important drawbacks. McKibben muscle showed substantial hysteresis and has a high threshold pressure, under which no contraction occurs. Also, the total displacement is limited to 20% to 30% of its initial length by the braided sleeving. The main cause for hysteresis in McKibben muscle is the Coulomb friction between the braided sleeving and the internal bladder.

Pleated PAM on the other hand has virtually no threshold pressure due to the expanding of muscle by unfolding the pleats. Also, hysteresis is reduced compared to McKibben muscle. Lastly, the contractions can be as large as 40% of the initial length, depending on the slenderness. The main cause for hysteresis in pleated PAM is due to the force-contraction characteristic.

2.2 Patient Type

This project focused on the pathological conditioms where foot drop may occur. The incidence of foot-drop is difficult to determine since it is a symptom rather than a disease; therefore numerous pathologies may present with foot-drop as a confounding factor. Essentially foot-drop is due to a disruption of the motor control pathway that occurs in the brain, peroneal nerve, spinal cord and/or muscle. Foot drop may occur in numerous pathological conditions such as stroke, spinal cord injury, multiple sclerosis, cerebral palsy, and neurological trauma from accident or surgical complication (Blaya & Herr, 2004, Chin et al., 2009). The major complications of the gait of a patient with foot drop are foot slap and toe drag, which will increased the risk of tripping.

2.3 Gait Cycle

2.3.1 Introduction

Gait analysis is important when developing a power assisted orthosis. In studies of human locomotion, a walking cycle is typically broken down into two phases; the stance phase (60%) and the swing phase (40%) as shown in Figure 2.3. The gait cycle for the right side begins with heel strike of the right foot. At this point, both feet are on the ground. This is known as the initial double support phase. This sub-phase of the gait cycle is also known as weight acceptance as the body weight is shifted to one leg. Forward advancement begins when the left foot leaves the ground (ie. left toe-off). During the single support phase of stance, the right leg supports the body weight while the left leg advances forward. When the left foot hits the ground, it is the beginning of a second double support phase. As the right leg comes off the ground (toe-off), the body transitions into swing phase. During this phase, the limb advances forward in preparation for the next contact with the ground (Gates, 2002).

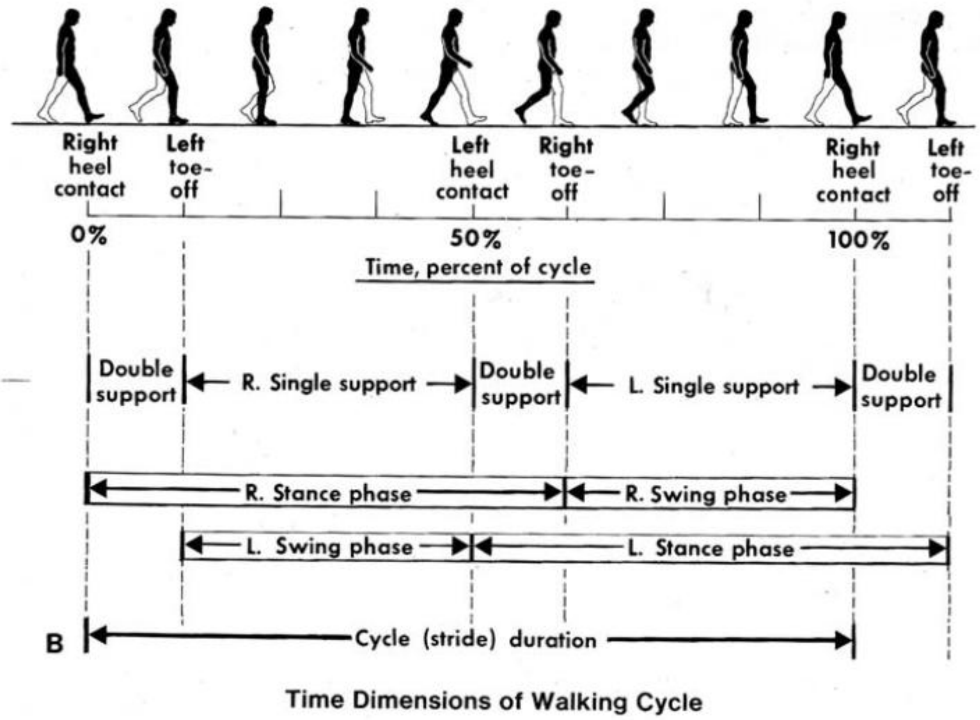


Figure 2.3: Gait Cycle. (Gates, 2002)

2.3.2 Ankle Characterization

The angle range of the motions of ankle during the dorsiflexion and plantarflexion is about -25 ° to 45 °. The angle velocity of motions is about 0 to 180 °/s (Li et al., 2008). Characterization of the ankle during walking was done in three parts. The first period of stance was controlled plantarflexion (CP). This phase began at foot strike (FS) and ended at the point where the minimum ankle position was reached. This position was referred to as foot flat (FF). The second period of stance was controlled dorsiflexion (CD), which lasted from FF until the point where the power became positive. The end of controlled dorsiflexion can be detected by identifying the occurrence of the maximum value of angular position. The third period that was studied was powered plantarflexion (PP). This began the instant the power became positive and lasted until the foot came off the ground (FO). Ankle angle, velocity, moment and power for one gait cycle of walking are shown in Figure 2.4 (Gates, 2002).

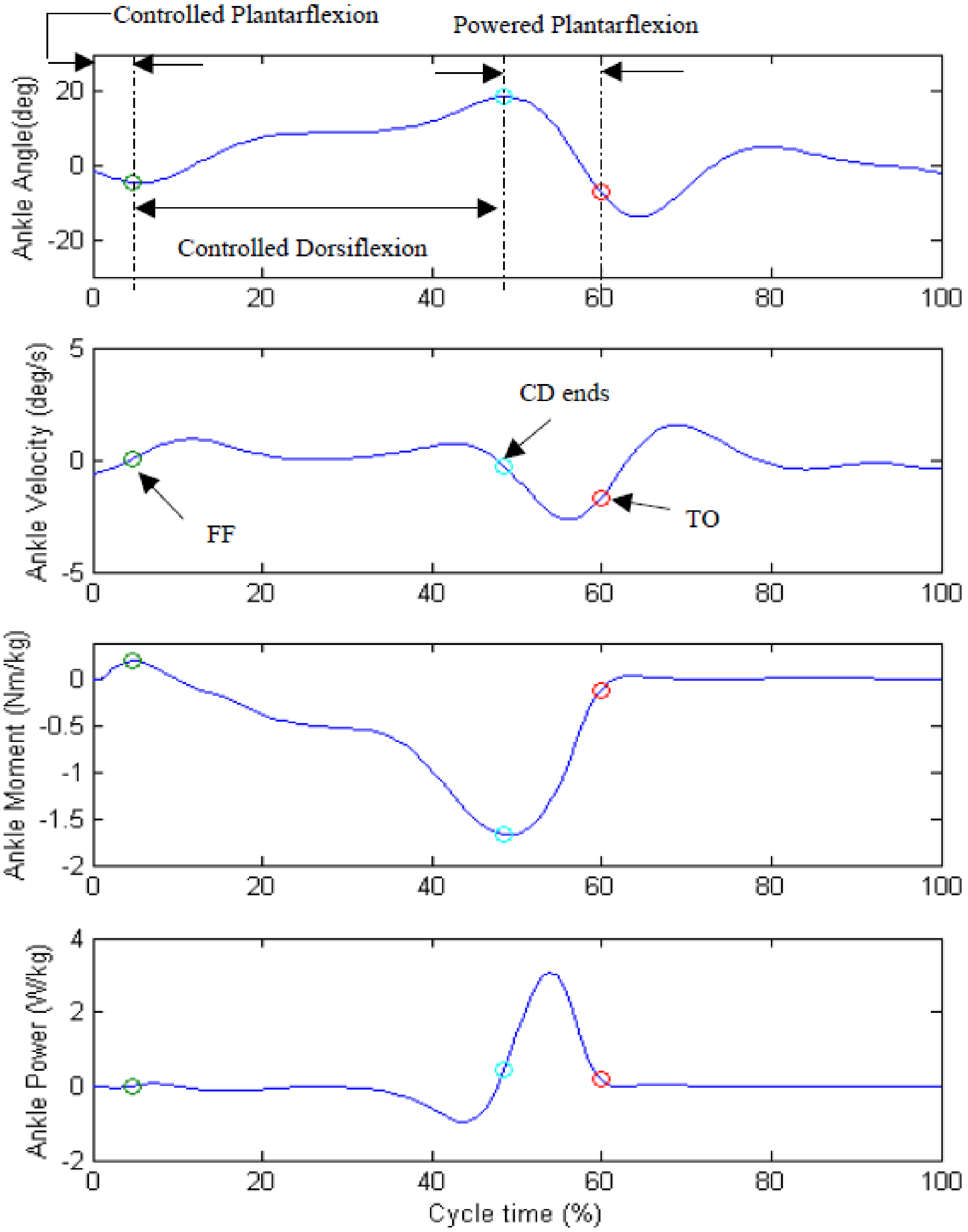


Figure 2.4: Ankle Angle, Velocity, Moment and Power for one Gait Cycle.

(Gates, 2002)

2.4 Sensor System

2.4.1 Electromyography (EMG)

Electromyography (EMG) signal is a signal produced by electrochemical activity involved in the motor unit activation. By using myoelectric control, a stable platform for acquisition of EMG signals produced during residual muscle contractions can be built. These signals are measured within the muscle tissue itself or at the external derma by means of either needle or surface electrodes. This type of sensing system has been used in wearable powered orthoses (Antonelli, Zobel & Giacomin, 2009, Farnsworth et al, 1992, Sawicki & Ferris, 2009).

The advantages of using EMG include this type of sensing system provides signals that the amplitude will increase as a function of increasing muscle force. Also, EMG provides a measure of command intention since the electrical activity build prior to actual muscular contraction (Antonelli et al, 2009). However, noises included in the signals make it difficult to identify the motions of the user accurately. It is difficult to obtain a reliable and consistent myoelectric signal due to surface electrode interface (Sawicki & Ferris, 2009). In addition, the motions of the user could not be identify correctly based on the observation of activities of several muscles only. This is due to the movement of a joint is actuated with the cooperation of many synergistic muscles (Hirata, Iwano, Tajika & Kosuge, 2009). Next, choosing the appropriate threshold and gain requires tuning (Sawicki & Ferris, 2009). Also, there is a possibility of mechanical detachment of the myoelectrodes due to body motion, electrical interference from nearby materials or devices, and interference caused by impedance changes of the skin (Antonelli et al, 2009).

2.4.2 Accelerometer

An accelerometer is an electromechanical device that will measure acceleration forces. These forces may be static, like the constant force of gravity pulling at feet, or they could be dynamic that caused by moving or vibrating the accelerometer. By measuring the amount of static acceleration due to gravity, the angle of the device is tilted at with respect to the earth can be calculated. Also, by sensing the amount of dynamic acceleration, the movement of the device can be analyzed.

Accelerometer has been used as the sensor system for wearable power assisted lower leg orthoses (Moreno, Brunetti, Rocon & Pons, 2008, Hirata et al., 2009). As mentioned by Hirata et al (2009), biological signals like EMG signal consist of a lot of noises, and measurement of a few muscles is not sufficient. To overcome these problems, non-biological signals could be use. The Wearable Walking Helper, a wearable walking support system developed by Hirata et al. calculates the support moment of the joints of the user by using an approximated human model of four-link open chain mechanism on the sagittal plane and it assists a part of the joint moment by the actuator of the wearable walking support system. The prototype of the system consists of knee orthosis, prismatic actuator and sensors.

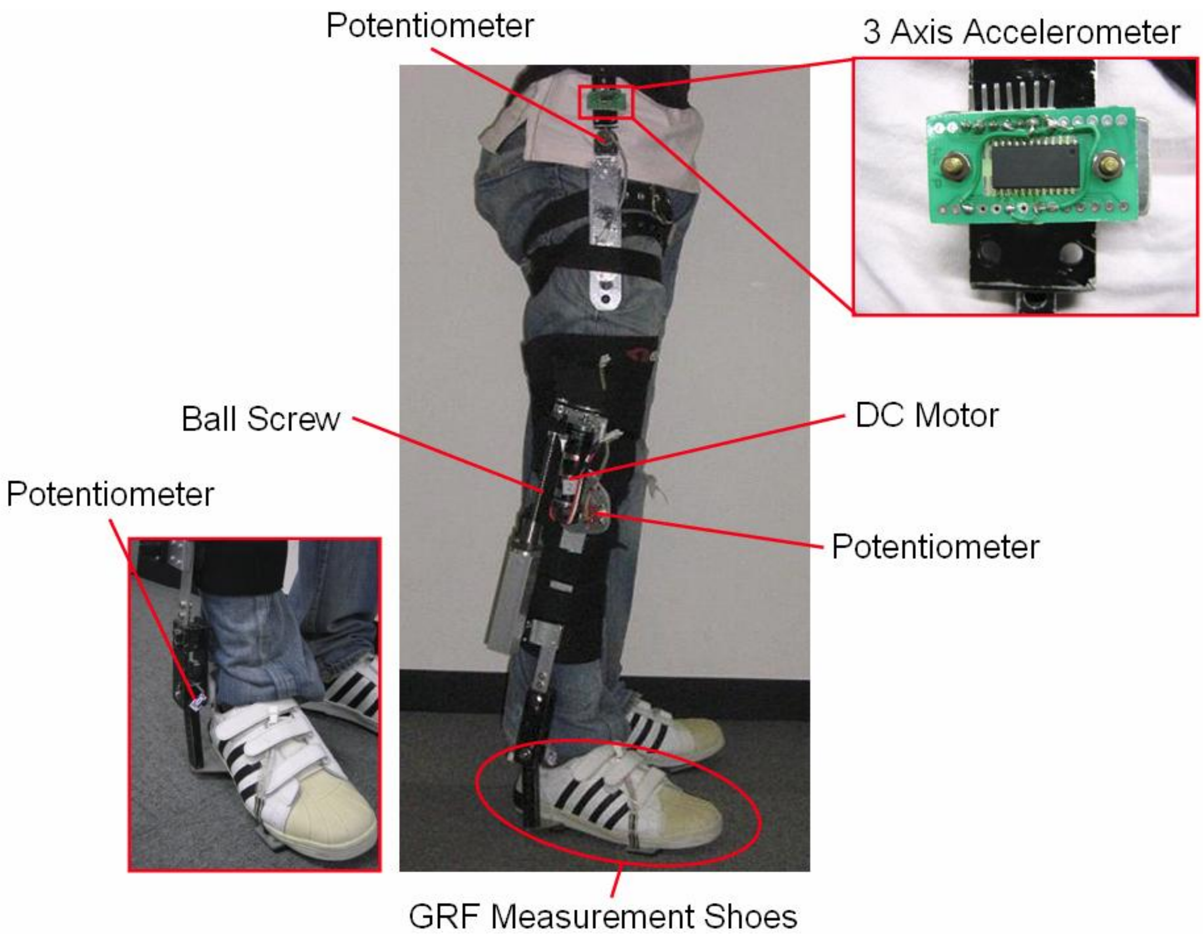


Figure 2.5: Wearable Walking Helper with Accelerometer (Hirata et al, 2009).

2.5 Robotic Rehabilitation Devices

Robotic rehabilitation devices for lower extremities can be divided into three categories (Allemand et al., 2009): stationary, mobile, and standalone.

2.5.1 Stationary Robotic Rehabilitation Devices

A treadmill based gait trainer associated with a body weight system (BWS) called Lokomat is developed and commercialised by Hocoma AG (Switzerland). In Lokomat, pelvic vertical movements, hip and knee joints are driven by orthoses linked to the treadmill frame with an appropriate mechanism. A similar device is developed by HealthSouth, which is called AutoAmbulator. The Pelvis Assist Manipulator (PAM) and Pneumatically Operated Gait Orthosis (POGO) project by Reinkensmeyer et al. (2006) on another hand is a system consists of two pneumatic driven robots to assist gait retraining. The LOPES robot (Lower-extremity Powered Exoskeleton) by Veneman et al. (2007) uses series elastic actuation. The motors fixed to the treadmill frame transmit the mechanical power via Bowden cables to the actuated joints (total of eight DOF). The GaitTrainer by Werner et al. (2002) and HapticWalker by Schmidt et al. (2007) replace the treadmill by two motorized footplates with BWS system. These end-effector machines allow different foot trajectories: walking, up/downstairs stepping.

Figure 2.6: Pelvis Assist Manipulator (PAM) and Pneumatically Operated Gait Orthosis (POGO) (Reinkensmeyer et al., 2006).

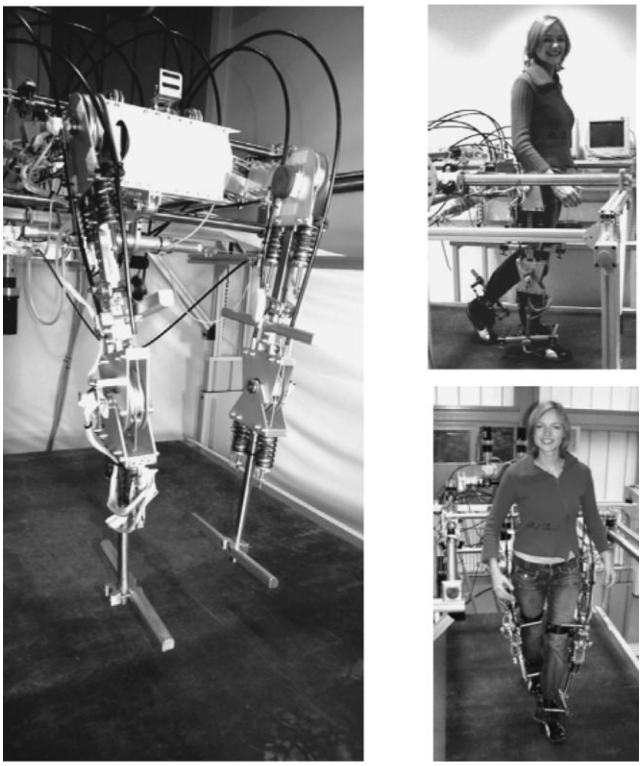


Figure 2.7: The prototype of LOPES (Veneman et al., 2007).

2.5.2 Mobile Robotic Rehabilitation Devices

Two mobile rehabilitation devices were reviewed, the KineAssist and also the WalkTrainer. The KineAssist is a mobile device for overground gait training developed by Peshkin et al. (2005). There is no leg orthosis, but assistance is provided at the pelvis as well as balance control. The BWS is integrated to the pelvic orthosis with a custom designed harness. The lower extremity must be managed by therapists if needed. The KineAssist has an omnidirectional mobile base allowing interesting movements such as sidestepping. The WalkTrainer by Allemand et al. (2009) on another hand is a locomotion training device with harness-based BWS system linked to the frame and mobile base is controlled by two motors driven at rear in a differential mode.

Figure 2.8: The WalkTrainer Prototype (Allemand et al., 2009).

2.5.3 Standalone Robotic Rehabilitative Devices

Standalone robotic rehabilitation devices that has been reviewed include Berkeley Exoskeleton BLEEX by Zoss, Kazerooni and Chu (2006), Hybrid Assistive Leg (HAL-5) by Kawamoto, Suwoong, Kanbe and Sankai (2003), RoboKnee by Pratt, Krupp, Morse and Collin (2004), and Quasi-Passive Exoskeleton by Walsh (2007). These exoskeletons assist and strengthen human performance. The movements are detected by sensors, for example EMG, force and position sensors. Also, BLEEX and HAL-5 can manage an extra payload besides their own weight.

Figure 2.9: BLEEX (Zoss et al., 2006).

CHAPTER 3

METHODOLOGY

3.1 Design Consideration

The wearable power assisted pneumatic based AFO consists of five main parts: 1) AFO, 2) frames, 3) PAMs, 4) actuation system, and 5) control system. Actuation system and control system will be discussed for both knee and ankle joint as they are closely related in this project.

There are few considerations for the design of AFO. Firstly, the device should be easy to put on and take off. Also, comfort of the patient is another consideration. Another important design consideration is the AFO should not restrict the movement of the subject's leg muscles from functioning to avoid muscle atrophy. Moreover, the frame of the AFO should be light weight and allow ankle movement within safety range. The PAMs should be able to produce sufficient strength and contraction length, and also can be fabricated easily. Also, another consideration is the cost to fabricate the PAMs. Next, the actuation system should provide sufficient air pressure without exceeds the limit of actuator to each PAM, whereas the control system should provide stimulation signal to each actuator in a control manner so that a normal gait cycle can be performed by the subject. The consideration of actuation system and control system are made for both knee orthosis and AFO. The conceptual designs are done with SolidWorks and Anim8or software and are shown in appendices.

3.2 Ankle-Foot Orthosis

The AFO is fabricated using orthopaedic casting tape, the material of which is fibreglass. Fibreglass cast is used by orthopaedic to immobilize fracture bone and had been used as mould for customised orthotic. To fabricate a brace that is firmly attached to the subject, the subject's lower leg is used as a mould. The fabrication steps are as below:

1) Wear two layers of orthopaedic stockinet on the moulding area. Draw the designated area where the orthopaedic casting tapes to be applied on the outer layer.
2) Open and immerse a roll of orthopaedic casting tape into room temperature water for five to ten seconds. Squeeze the roll for few times while in the water.
3) Apply the orthopaedic casting tape on designated area, overlapped each layers by at least half of the width. Each area needs to have at least four layers of tapes.
4) Rub the tape between layers and on the surface to improve lamination and layer adhesion.
5) Wait for about 5 minutes for the orthopaedic casting tape to become sufficiently rigid to prevent further moulding. Heat will be generated by resin during the hardening process.
6) Remove the cast from the orthopaedic stockinet.
7) Wait for another 30 minutes to let the heat generated by resin fully dissipated and to apply pressure on the cast.
8) Cut the cast into designated shape.

The design of the orthosis is to allow muscle in leg to function normally. The gastrocnemius and soleus muscle (Figure 3.1) will not be covered by the fibreglass cast. Gastrocnemius and soleus muscle help to perform plantar flexes foot at ankle joint. Gastrocnemius muscle also help to flex leg at knee joint (Tortora & Derrickson, 2007). On another hand, to ease the patient to apply and remove the AFO, it is designed with Velcro strap. By using Velcro strap, the AFO is also fit the size of patient in a certain range according to anthropometric parameter (Table 3.1) by

Plagenhoef, Evans and Abdelnour (1983). Also, fitting with strap also help the airflow to the leg, which will help to prevent sweating and odour. Figure 3.2 shows the AFO designed.

Table 3.1: Anthropometric Parameter.

SEGMENTS	**MALES**	**FEMALES**
LOWER LEG	24.70	25.70
FOOT	4.25	4.25
*Segments lengths expressed in percentages of total body heights.		

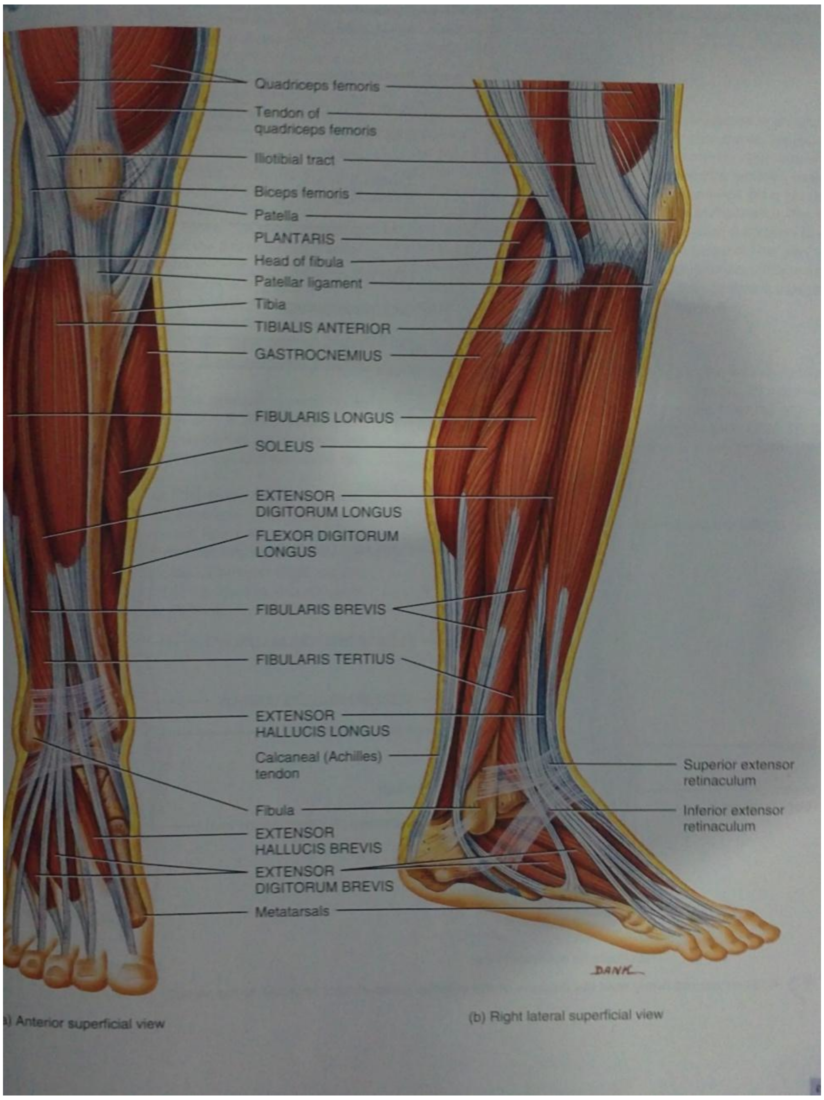


Figure 3.1: Anatomy of Lower Leg. (Tortora & Derrickson 2007)

Figure 3.2: Ankle Foot Orthosis Using Orthopaedic Casting Tape.

3.3 The Frames

Aluminium was chosen as the main material for the frame due to its light weight and high strength properties. The main skeleton of the frame is made by 1” × 2” aluminium bar whereas aluminium plate with 1mm thickness is chosen for the mounting of PAMs. The frame is designed to hold the orthoses (thigh and lower leg) together. The attachment is designed in such that the brace for lower leg is adjustable to fit patient with height within 1.6m to 1.8m. The fitting can be done by adjusting the attachment of the brace to the frame. Figure 3.3 and Figure 3.4 show two different attachment of the AFO on the frame. The frame also serves as the mounting for the antagonist pair pleated PAM as shown in Figure 3.5. The aluminium plates are attached to the aluminium bar using rivet.

A hinge joint that has one degree of freedom is designed to replace the ankle joint function. The hinge joint angle is limited to -15 ° to 10 ° to match the ankle movement during gait cycle and also for safety purpose by using mechanical stopper as shown in Figure 3.6 and Figure 3.7. On another hand, stainless steel plate is chosen for the foot plate as a stronger material was needed for the foot plate to carry the body weight. The foot plate is designed to fit any sizes of shoes and attached to frame by using bolts and nuts. The foot plates are made of 2 smaller pieces of stainless steel instead of a big plate is due to two reasons. Firstly is to reduce the weight of the prototype. To create plantarflexion and dorsiflexion at ankle joint, the actuators have to actuate at two spot only, the heel and mid foot. Secondly is to reduce the noise made by the foot plate. The noise create by two small piece of stainless steel is significantly lower than a large steel plate. The design of frame is shown in Figure 3.8 to Figure 3.10.

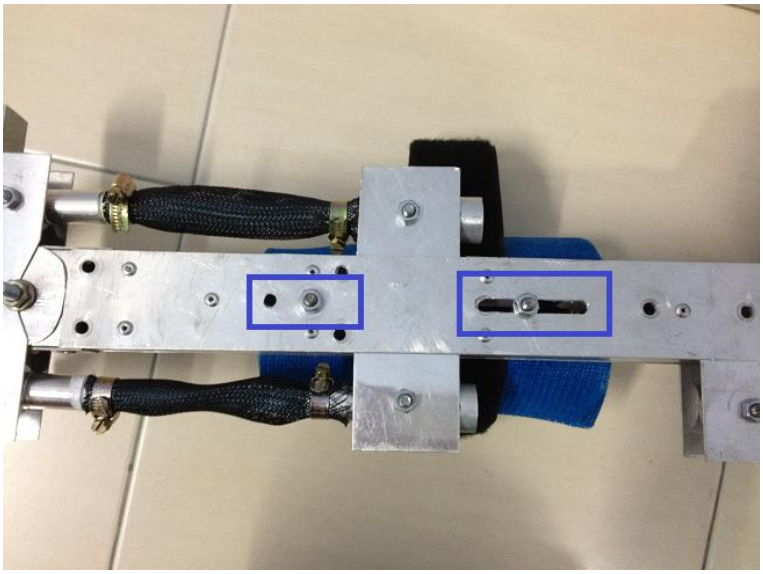


Figure 3.3: Attachment of AFO at Higher Position.

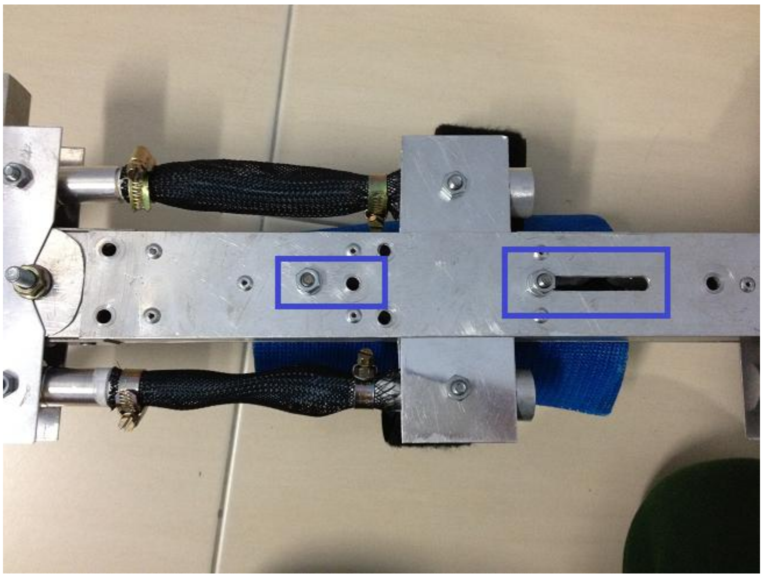


Figure 3.4: Attachment of AFO at Lower Position.

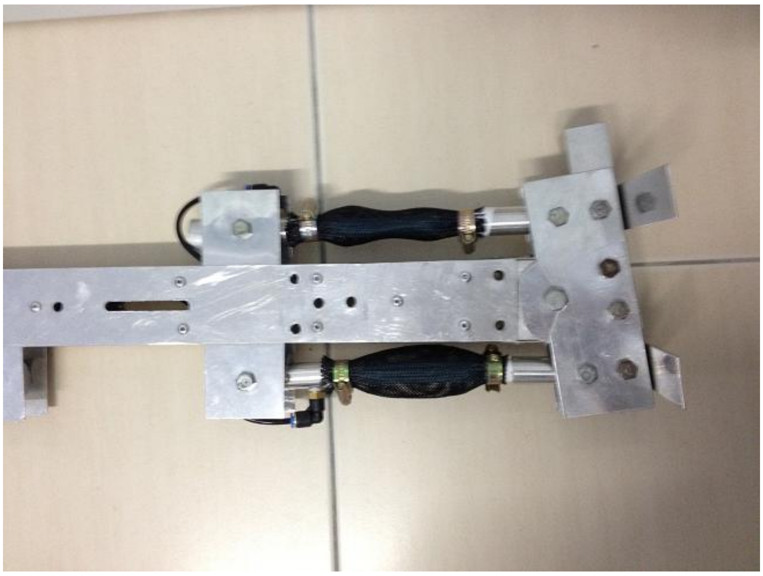


Figure 3.5: Mounting of McKibben PAM on the Frame.

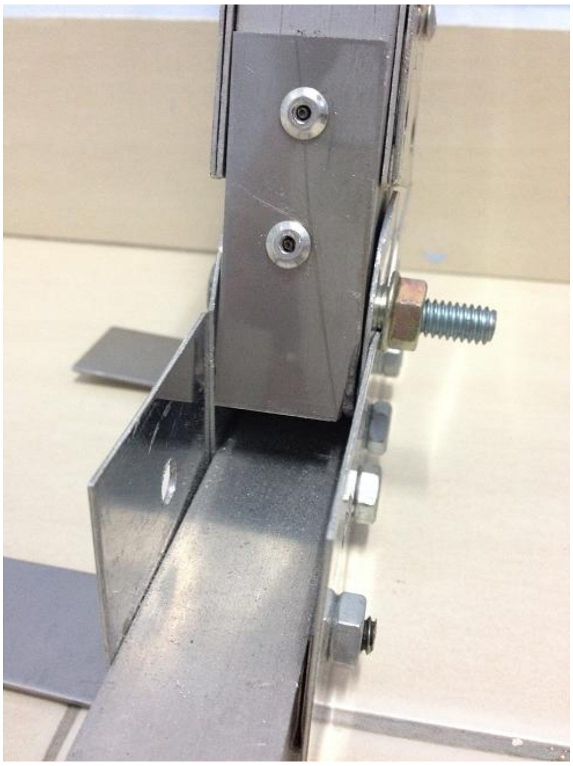


Figure 3.6: Mechanical Stopper (Front) to Prevent Hyperdorsiflexion.

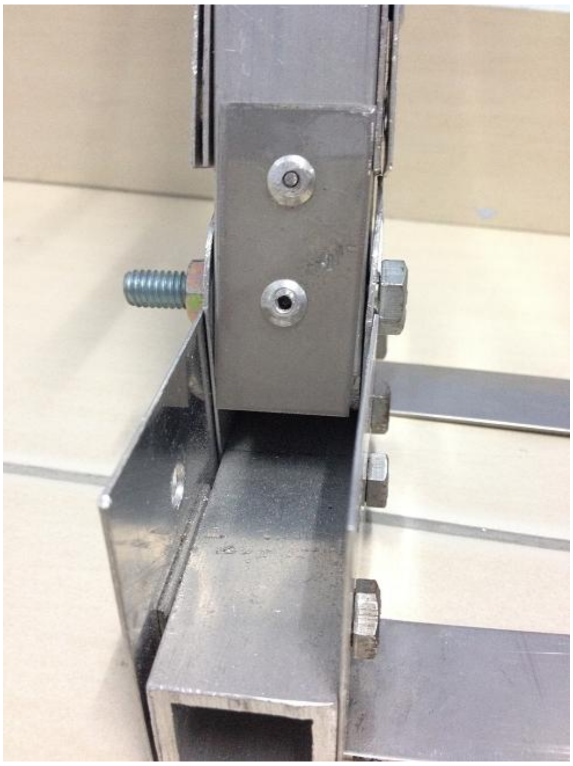


Figure 3.7: Mechanical Stopper (Back) to Prevent Hyperplantarflexion.

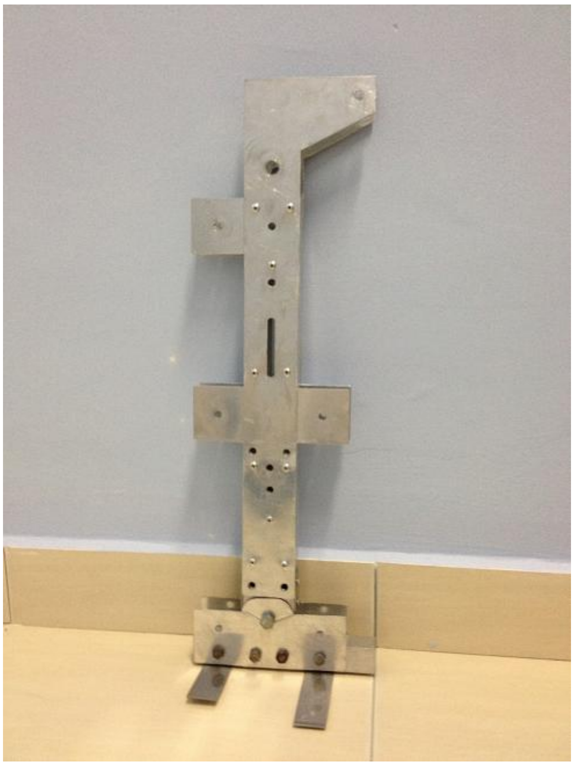


Figure 3.8: Right View of the Frame.

Figure 3.9: Front View of the Frame.

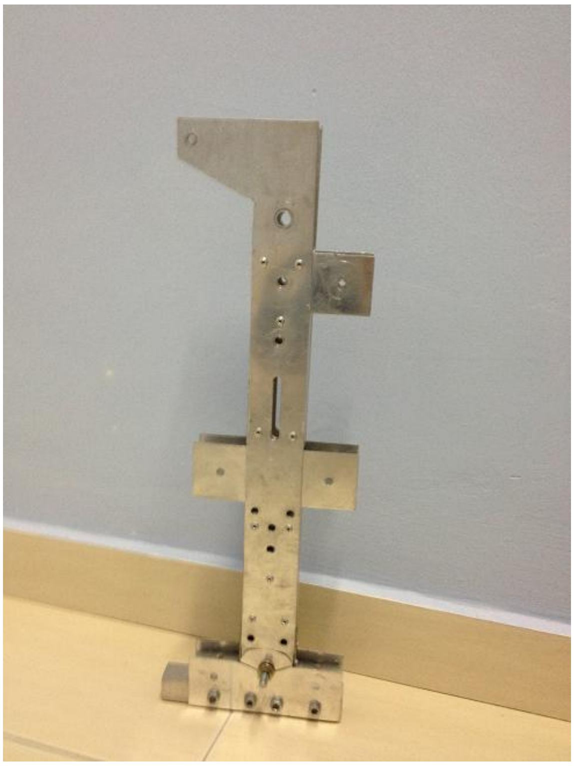


Figure 3.10: Left View of the Frame.

3.4 Pneumatic Artificial Muscles

After considering the cost of material and ease of fabrication, McKibben PAMs was chosen for the project. The McKibben PAMs can be fabricated by using a few simple materials as shown in Table 3.2.

Table 3.2: Material required for fabrication of McKibben PAMs.

	PAMs end cap

	4mm tube ×1/8” BSPT thread- Swivel elbow adaptor
	Braided nylon sleeve
	Latex glove
	Hose clip
	White tape

The muscle end cap is fabricated using CNC turning machine. The raw material of a muscle end cap is aluminium rod. A 40mm deep tunnel with 6mm diameter was drilled from the end of the muscle end cap, and a connection for the swivel elbow adaptor is made using BSPT 1/8” taper. Two more holes were drilled at the top of the muscle end cap for mounting purpose. Figure 3.11 and Figure 3.12 show the specifications of the muscle end cap.

Figure 3.11: Specification of Muscle End Cap (Perspective View).

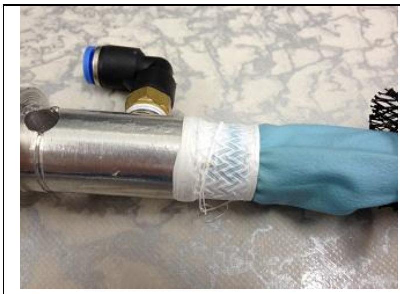
Figure 3.12: Specification of Muscle End Cap (Side View).

3.4.1 Fabrication of McKibben PAMs

The fabrication process of the McKibben PAMs are summarise in the table below:

Table 3.3: Fabrication process of McKibben PAMs.

	Cut the braided nylon sleeve into desired length.
	Burn the end of the braided nylon sleeve to prevent fraying.
	Screw the 4mm tube ×1/8” BSPT thread- Swivel elbow adaptor at the 1/8” BSPT hole.
	Cut a finger of latex glove.
	Bandage the attachment area for latex glove with white tape.

	Attach the latex glove onto end cap and bandage another layers of white tape to prevent slipping and air leak.
	Slip in the braided nylon sleeve and clip it with a hose clip.

3.5 Actuation system

Figure 3.13 shows the schematic diagram of the actuation system that is designed for the project. An air compressor is used to generate air pressure to operate the PAMs. The air generated by air compressor will firstly flow through an air service unit which contain water filter, regulator and pressure gauge. Then, the air is separate into four ways using three 4mm tee pneumatic connector. Each tube from the tee connector is connected to a 5/2 way solenoid valve. Before the air from the solenoid valve is feed to PAM, the velocity of the air flow is regulated by using one-way flow control valve. Figure 3.14 and Figure 3.15 shows the different part of actual actuation system of the project. 5/2 way solenoid valve is chosen for the actuation system is due to its flexibility on function. Comparing to 3/2 way solenoid valve, 5/2 way solenoid valve has two more port on each chamber. Hence, a 5/2 way valve is able to function as a 3/2way valve when the extra two port is not needed by closing the ports.

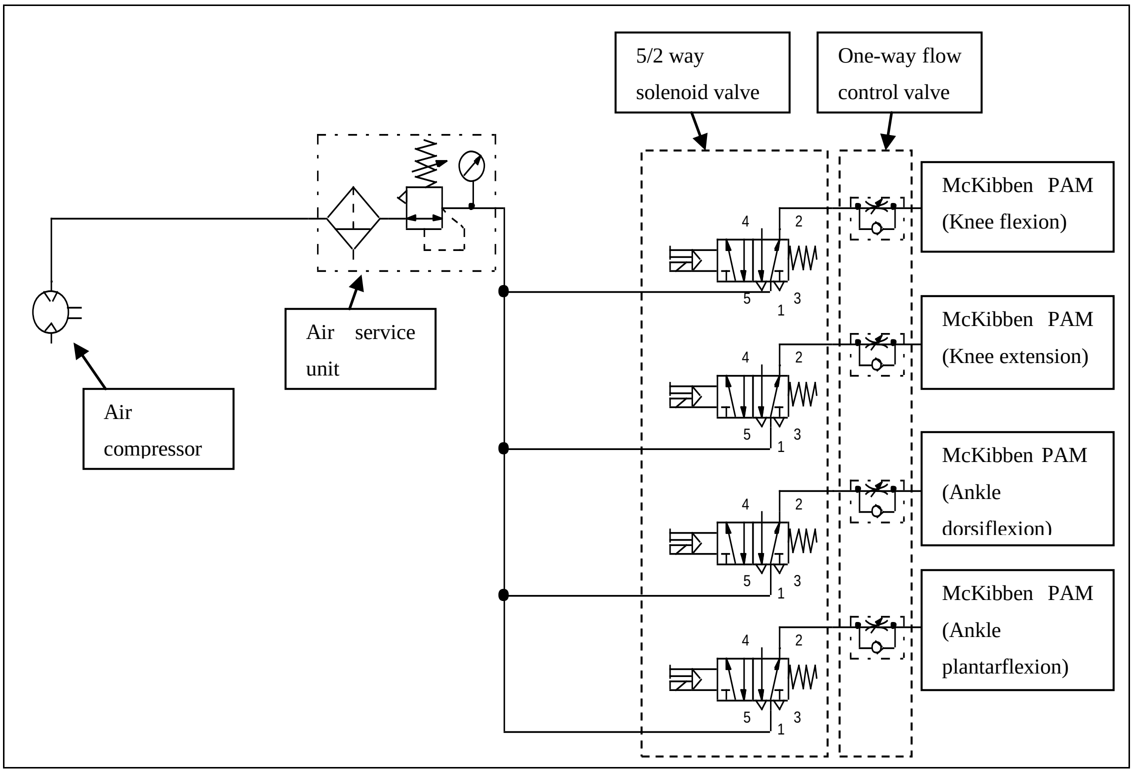


Figure 3.13: Schematic Diagram of Actuation System.

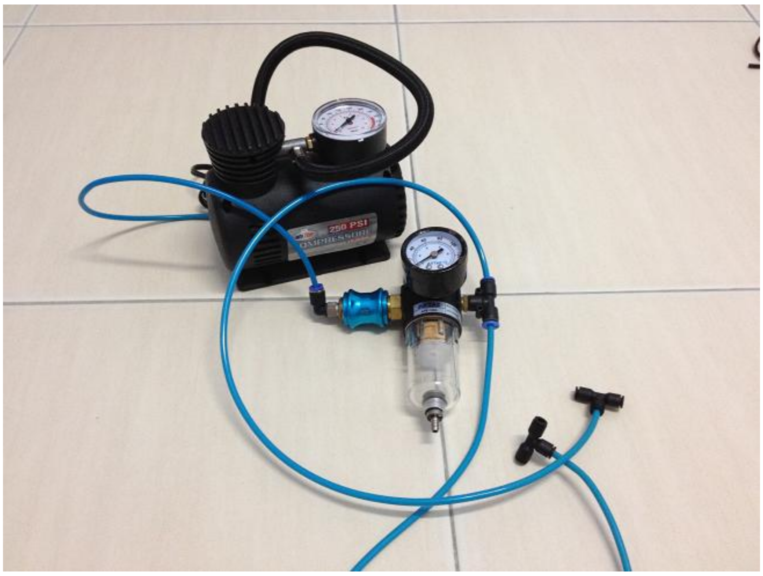


Figure 3.14: Air Compressor and Air Service Unit.

Figure 3.15: 5/2 Way Solenoid Valve, One-way Flow Control Valve and McKibben PAM.

The movement of joint is determined by the contraction of corresponding PAM. The relationship between inclination angle of each joint and McKibben PAMs' displacement can be expressed in a formula using trigonometry method.

$$\Delta x = - \Delta z \tag{3.1}$$

$$\Theta = 90° - tan\frac{w}{v} - \ cos^{-1}\left(\frac{x^2 + y^2 - w^2 - v^2}{2xy}\right) \tag{3.2}$$

where

w, v, y = length of respective part, cm

x, z = length of muscle, cm

Ɵ = inclination angle, °

Figure 3.16: The Design of Muscle Configuration.

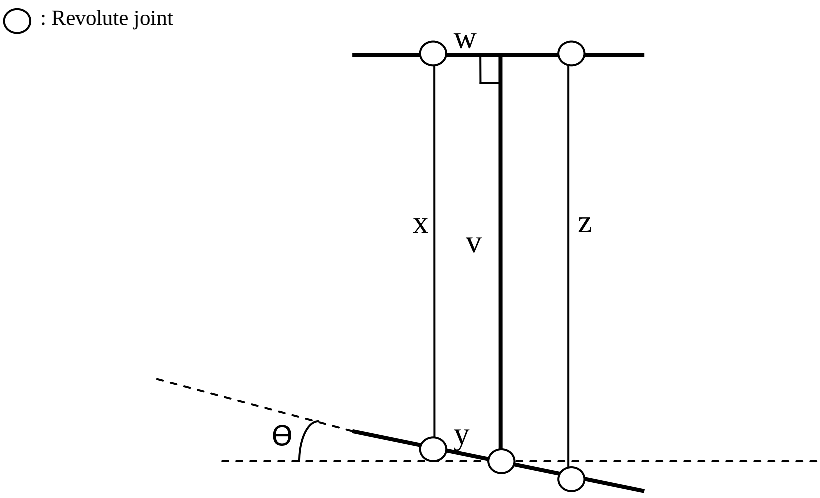


Figure 3.17: Mathematical Model of Muscle Configuration.

3.6 Control System

The Figure 3.18 shows the schematic diagram of the overall control system for this project. A microcontroller is used to control the actuation system by gaining feedback from the accelerometer and also start/ reset button. The accelerometer will sense the movement of the thigh segment to initiate the control system. PIC18F4520 was chosen for this project. The microcontroller will send a 5V impulse to the relays which control the 24V power input to the solenoid valves. Other outputs of the microcontroller include LED blinking and LCD display.

To ease the movement of the subject while wearing the prototype, some component of the control system is built on an acrylic board, including a Cytron SK40C PCB board for PIC microcontroller, a PCB board for relays and voltage regulators, and sockets for batteries and solenoid valves. The acrylic board, batteries, and solenoid valves are mounted into a backpack.

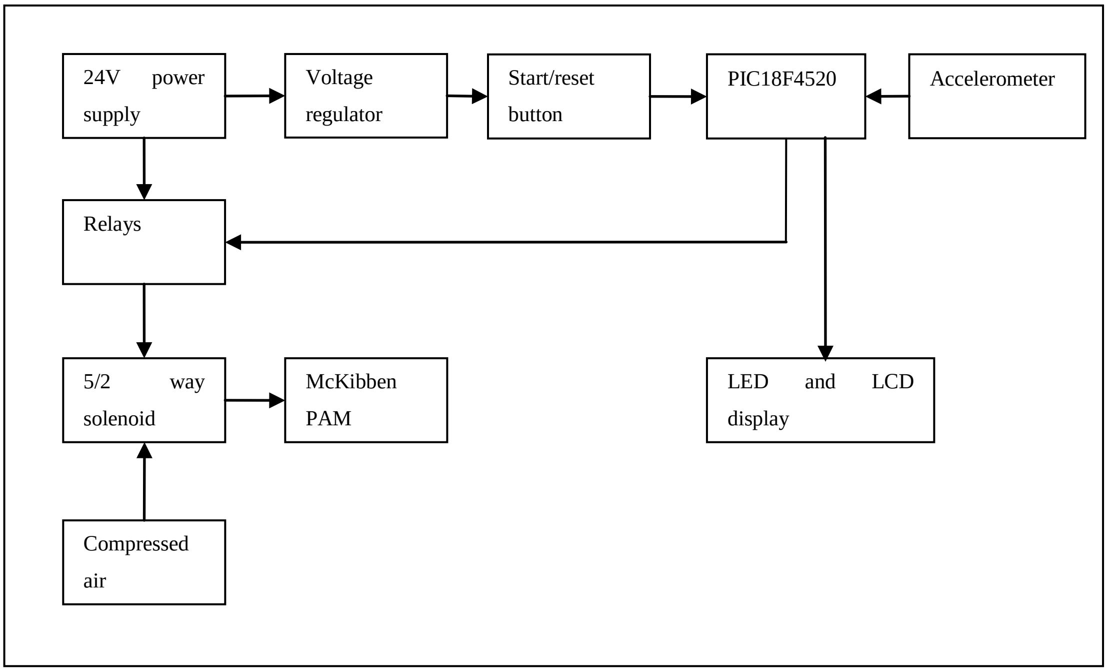


Figure 3.18: Schematic Diagram of Control System

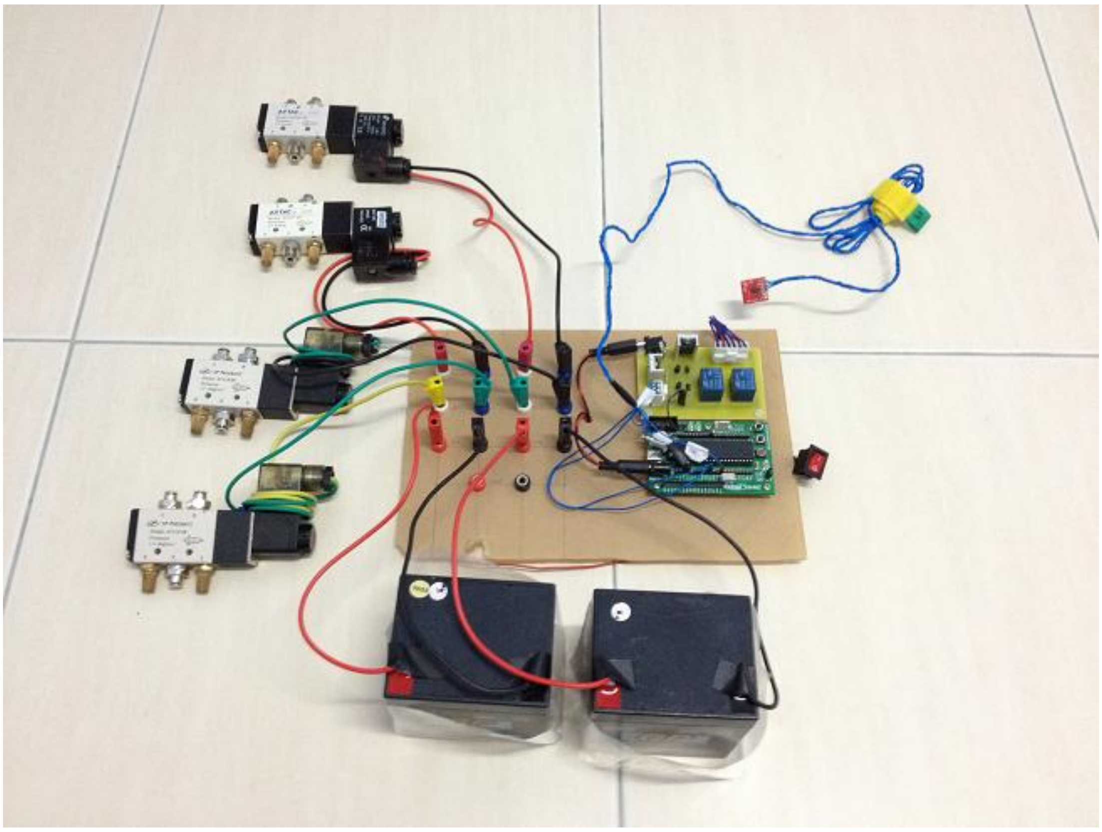


Figure 3.19: The Actual Control System of the Project.

3.7 The Overall System of the Prototype

Figure 3.20 and Figure 3.21 show the overall system of the prototype including AFO, frames, PAMs, actuation system, and control system. An air compressor, an air service unit, four 5/2 way solenoid valves, two 12V sealed lead acid batteries, an acrylic board with components and the knee-ankle-foot orthosis (KAFO) are shown in Figure 3.20, whereas all components except the KAFO are mount into a backpack in Figure 3.21.

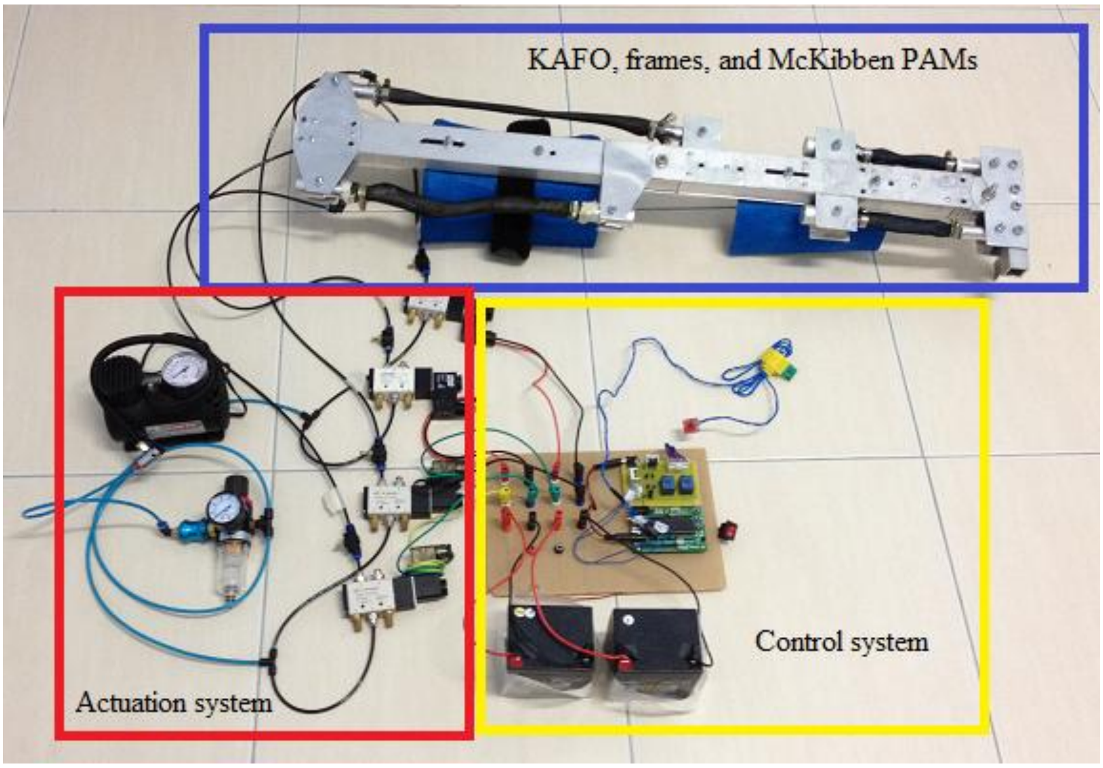


Figure 3.20: Overall system of Wearable Power Assisted Pneumatic-based Knee-ankle-foot Orthosis (KAFO) Prototype.

Figure 3.21: The KAFO with Other Components Mounted in a Backpack.

CHAPTER 4

RESULTS AND DISCUSSIONS

4.1 McKibben Pneumatic Artificial Muscles

Experiment is conducted to test the properties of the muscle to actuate ankle joint movement. The experiment setup was to determine the contraction length of the McKibben PAMs against different load. The setup of the experiment was hanging the McKibben vertically and apply different load below the McKibben PAMs. Load of 1kg, 2kg, 3kg, 3.5kg and 4kg was chosen for the experiment. The load was limited to 4 kg due to detachment or explosion of the inner tube which is the finger part of a latex glove often occur when the pressure exceeds 5kg. Also, the pressure supply to the McKibben PAMs is limited to 20psi to prevent detachment or explosion of the inner tube. The result of the experiment wad concluded in Table 4.1 and Figure 4.1.

Table 4.1: Result of McKibben PAMs experiment

Load (kg)	Length of PAM without actuation (cm)				Length of PAM with actuation (cm)			
	1st try	2nd try	3rd try	Average	1st try	2nd try	3rd try	Average
1.0	26.4	26.5	26.5	26.5	22.9	23.0	23.0	23.0
2.0	26.6	26.6	26.6	26.6	23.1	23.1	23.2	23.1
3.0	26.6	26.6	26.6	26.6	23.4	23.5	23.5	23.5
3.5	26.6	26.6	26.6	26.6	23.6	23.6	23.7	23.6
4.0	26.6	26.6	26.7	26.6	23.6	23.7	23.7	23.7

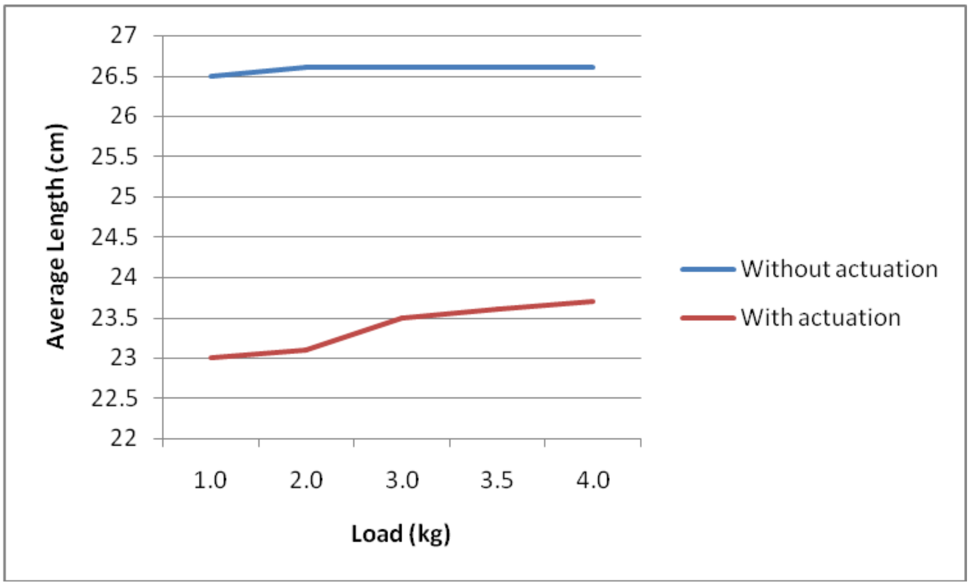


Figure 4.1: Graph of Average Length versus Load

From the table and graph, the length of McKibben PAMs without actuation did not differ much, which is preliminary limited by the maximum length of braided nylon sleeve. The length of the McKibben PAMs with actuation has shown a non-linear characteristic of contraction as the loading gradually increase from 1kg to 4kg. However, the McKibben PAMs were able to overcome the gravitational force of 4kg with 20psi pressure exerted in it. The contraction length of the McKibben PAMs are within the range of 2.9cm to 3.5cm, which are more than enough to create an inclination angle of -15 °to 10 °.

There are two limitations can be found in the McKibben PAMs. Firstly, the inner tube of the McKibben PAMs was tending to detach from the muscle end cap or explode when the load exceeds 5kg. The detachment might due to the clamping force of the hose clip are smaller than the force exerted. To solve this problem, a tighter clamping is used. However, this solution will lead to another limitation, where the explosions of the inner tube are mostly happened at the end of the inner tube where near the attachment to the muscle end cap. It is suggested that the clamping force of the hose clip had damaged the inner tube, which is made of latex. Hence, the clamping cannot be too tight which will lead to more frequent explode of inner tube.

Next, the performance of McKibben PAMs while attached to the frame during gait cycle was determined. The length of McKibben PAMs after contraction for both dorsiflexion and plantarflexion is recorded in Table 4.2. When no movement, the ankle joint is in anatomical position which perform a 90 ° angle between foot segment and lower leg segment.

Table 4.2: Length of McKibben PAMs during gait cycle.

Movement	Length of plantarflexion muscle (cm)	Length of dorsiflexion muscle (cm)
No movement	25	25
Dorsiflexion of 15 °	27	23
Plantarflexion of 10 °	24	26

From the collected data and through observation, the contractions of McKibben PAMs are sufficient to create the inclination angle desired. From the first experiment, the contraction length of the muscle were 2.9cm to 3.5cm, whereas to perform dorsiflexion of 15 ° and plantarflexion of 10 ° required only contraction of 2cm and 1cm, respectively. Further inclination of the ankle joint is restricted by the mechanical stopper, which functions as a safety configuration and also aid in adjusting the gait pattern.

4.2 Performance of the Prototype

The analysis of the overall performance of the prototype is done with the full prototype, which is a power assisted pneumatic-based KAFO. A walking gait which is near to normal gait pattern can be performed by the subject while wearing the prototype. The gait pattern at sagittal plane is recorded into a video to compare with a normal gait pattern. The comparison of the gait pattern is shown in Figure 4.2.

Figure 4.2: Comparison of normal walking gait with the actual walking gait performed by the subject.

By comparing the actual walking gait performed by the subject, shorter single limb stance phase on left leg, which is the leg wearing the prototype was observed. As the result, the right steps are also shorter in compare with left steps. Also, the contraction time of each muscle is independent from each other. Although it is set in the programming in control system, some adjustment still needed. These adjustments can be done by adjusting the air flow velocity into each McKibben PAM using the one-way flow control valve.

Another testing is done to determine the vertical ground reaction force (VGRF) during gait cycle using treadmill with embedded force plate. Two experiments were done, first with a subject wearing the prototype without actuation and second with the subject wearing the prototype with actuation. Figure 4.3 to Figure 4.5 show the VRGF versus time graph of a sample step from each leg for each experiment.

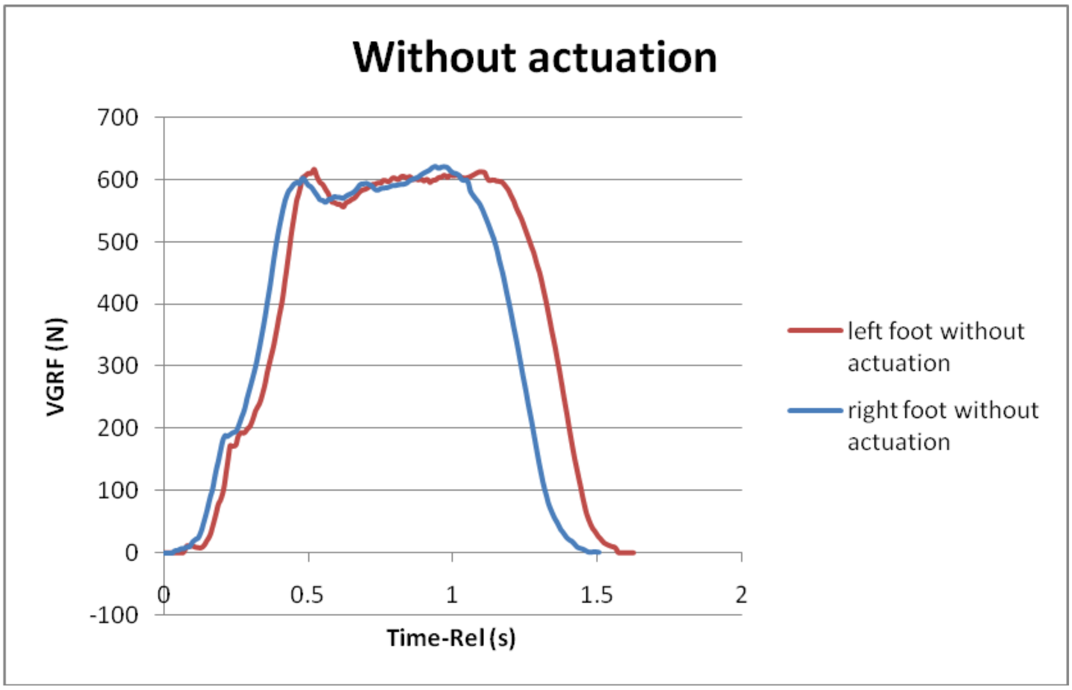


Figure 4.3: Graph of VGRF versus Relative Time for Experiment without Actuation

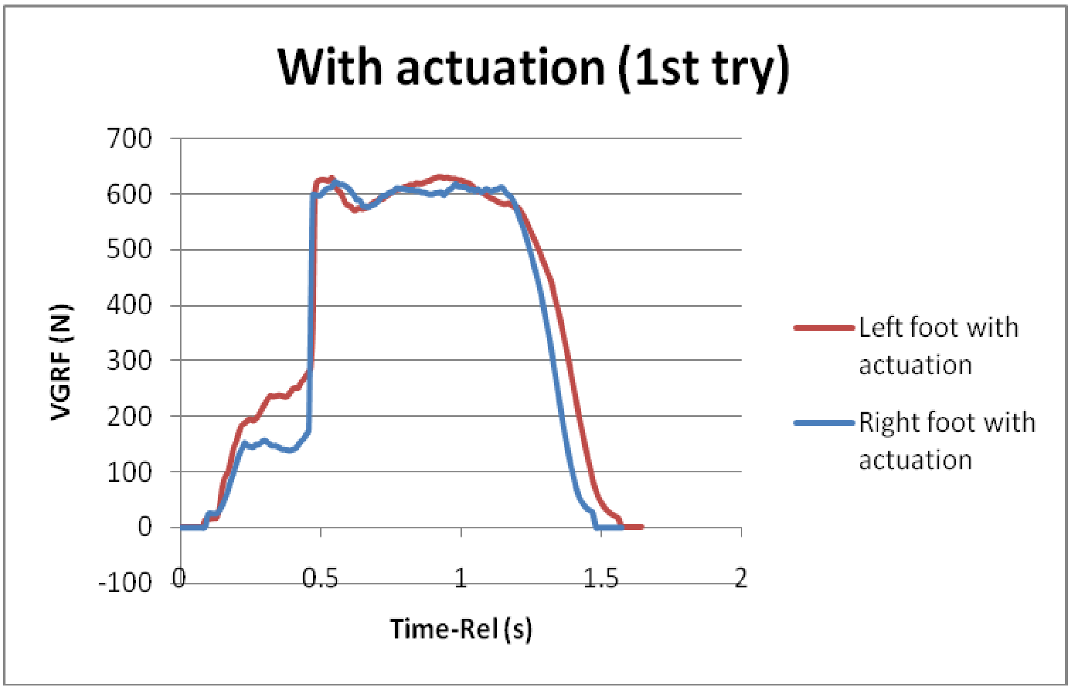


Figure 4.4: Graph of VGRF versus Relative Time for Experiment with Actuation (1st)

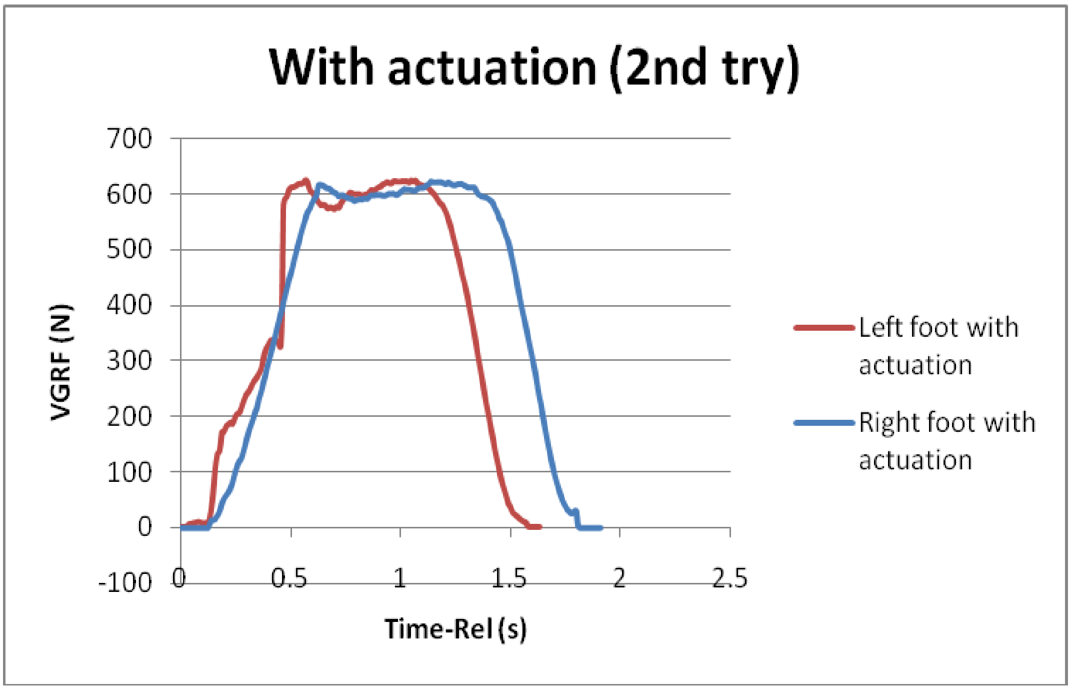


Figure 4.5: Graph of VGRF versus Relative Time for Experiment with Actuation (2nd)

From Figure 4.3, longer left stance phase was observed from the gait cycle while wearing the prototype without actuation. This might due to the weight of the prototype, which is wore on the left leg require more strength from the subject to raise his left leg. Figure 4.4 shows the right foot had imbalance during shifting the force from heel to toe. This might due to the subject was not used to wear the prototype. Figure 4.5 shows that the right stance phase is longer than left stance phase. This might due to the setting of the programming and also the setting of the one-way flow control valve.

During the experiment, imbalance often occurred while the subject wearing the prototype with actuation. Also, the small compressor shown in Figure 3.19 was not able to supply sufficient force to all four McKibben PAMs during walking. In both gait analysis experiment, the small compressor was replaced with a larger compressor with air tank that is able to store 100psi of air pressure. However, the air tank was too heavy that it cannot be carried by the subject. As the result, the mobility of the prototype is limited to the length of the hose from the air tank, which is five metres.

CHAPTER 5

CONCLUSION AND RECOMMENDATIONS

5.1 Conclusion

This project aims to design and develop a wearable power assisted pneumatic-based ankle-foot orthosis (AFO) at low cost to aid patient with lower extremities problem. The project also aims to aid patients during rehabilitation process. To design a system and hardware that fulfil the aims above, the gait cycle of human must be studied.

A wearable device has to be mobile. However, due to insufficient air supply from a small compressor, an air tank has to be used. A patient with lower extremities problem, or even normal person was unable to carry the weight of the air tank, which hence limit the mobility of the power assisted AFO. To solve this problem is to get an air compressor with high power but in small size, but that will exceed the budget for this project. Another solution is let the subject walk on a treadmill with the air tank located next to the treadmill and connected to the system with a hose. Although this will limit the movement of the subject on a treadmill, it is considerably sufficient for a patient that is undergoing rehabilitation process. Testing result and observation during the experiment suggested that this project is suitable for rehabilitation process which aims for patient to regain walking ability. Force generated by the McKibben PAMs is sufficient and the gait pattern performs by the subject wearing the prototype match the human gait pattern.

5.2 Future Work

There are a few improvement could be suggested for this project. Some drawbacks are identified in the current prototype, including the balancing of the subject while wearing the prototype, the mobility of the system and also lack of feedback from the device.

First improvement that can be done to this project are the produce of another power assisted KAFO for right leg. The imbalance experienced by the subject is due to the imbalance weight distribution at lower body and it takes time for the subject to get used with the system. With another power assisted KAFO on the right leg, the distribution of weight is more balance compare to just one leg wearing the power assisted KAFO. Also, we also suggest adding hip joint actuation to the system so that the gait pattern can be more similar to the actual human gait pattern.

Another improvement that can be done is adding more feedback system to the control system. Feedback systems that can be added including the recording of inclination angle of each joint using accelerometer. The record can be used as data analysis for medical practitioner or physiotherapist to understand the condition of the patient. Another feedback system that can be added is the detection of floor contact of foot. By detecting the floor contact, time of stance phase of each leg can be controlled.

Also, a compact compressor with high power can also be added into the actuation system. Current compressor is not capable to generate sufficient air pressure rapidly to provide equal air pressure to each McKibben PAMs, which forced us to replace it with a larger compressor with air tank. With all improvement added, the proposed wearable power assisted pneumatic-based KAFO, which is wearable by patients to aid in their daily life and rehabilitation can be produced.

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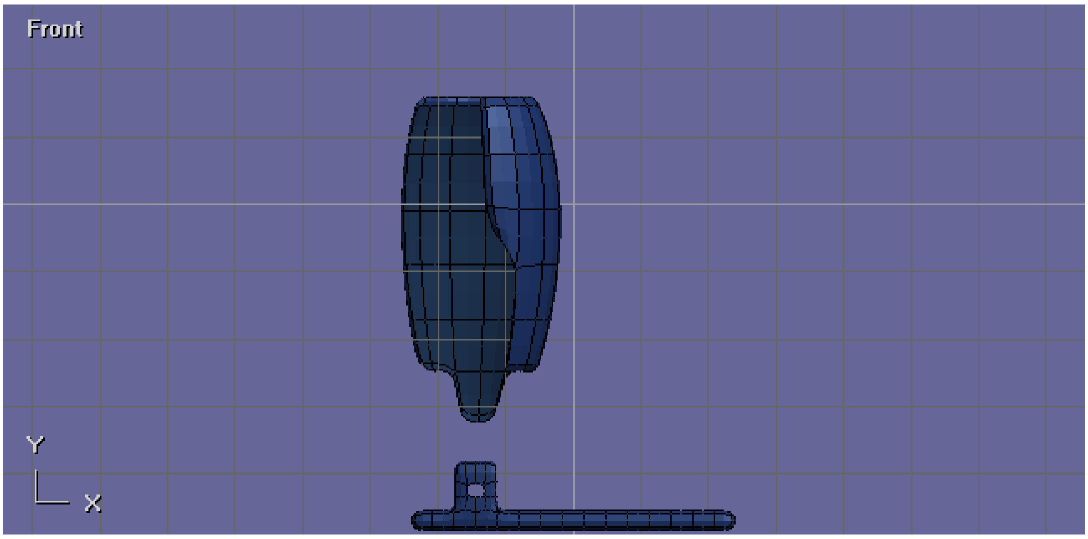
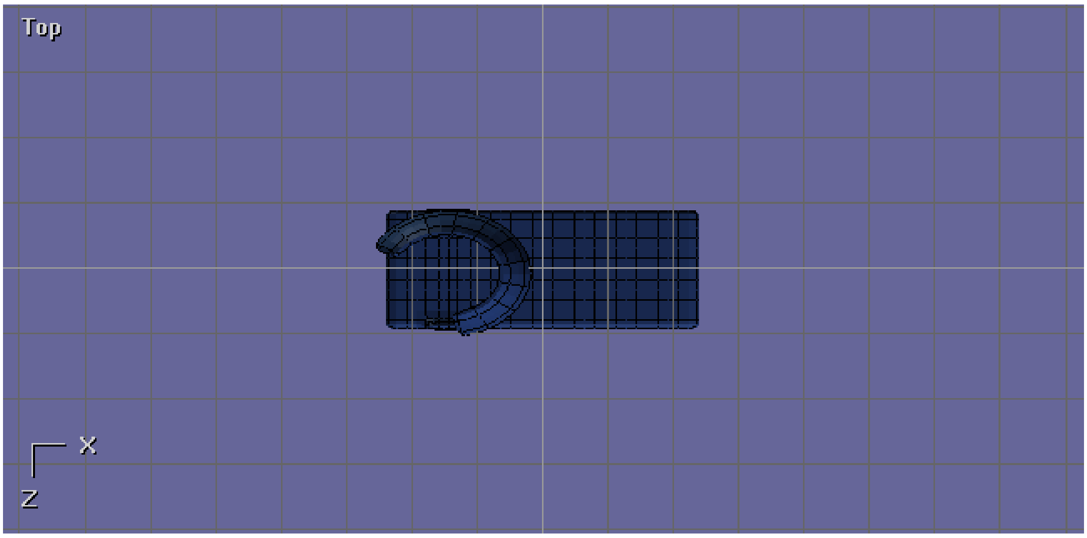
APPENDICES

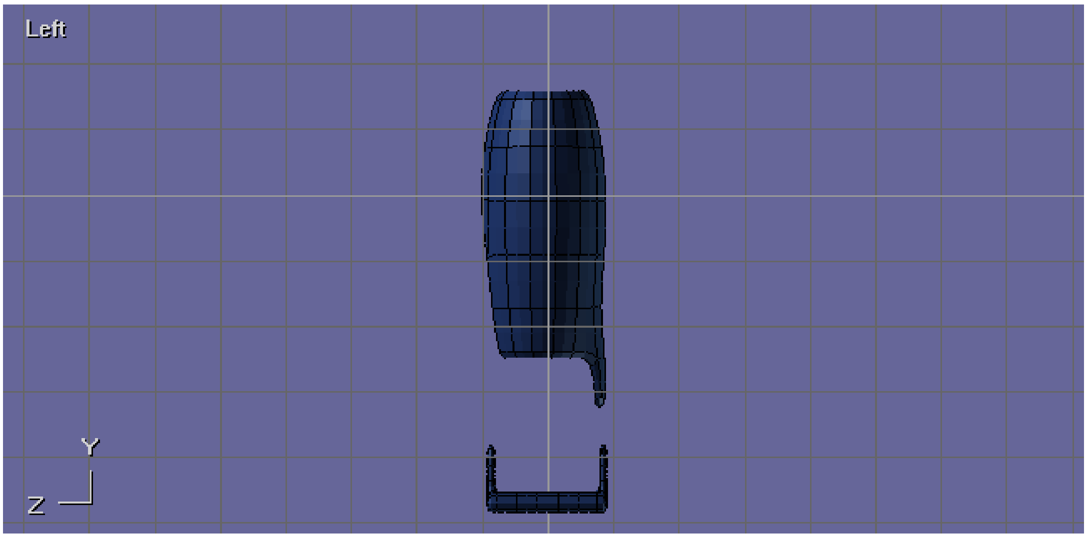
APPENDIX A: Conceptual Design Using Anim8or



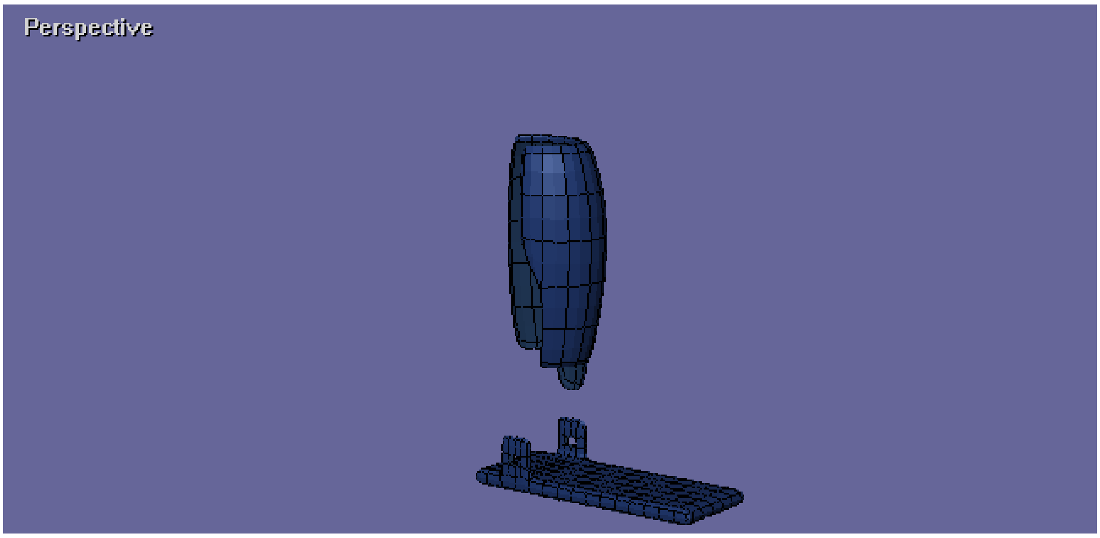
Conceptual Design of AFO (Front View)



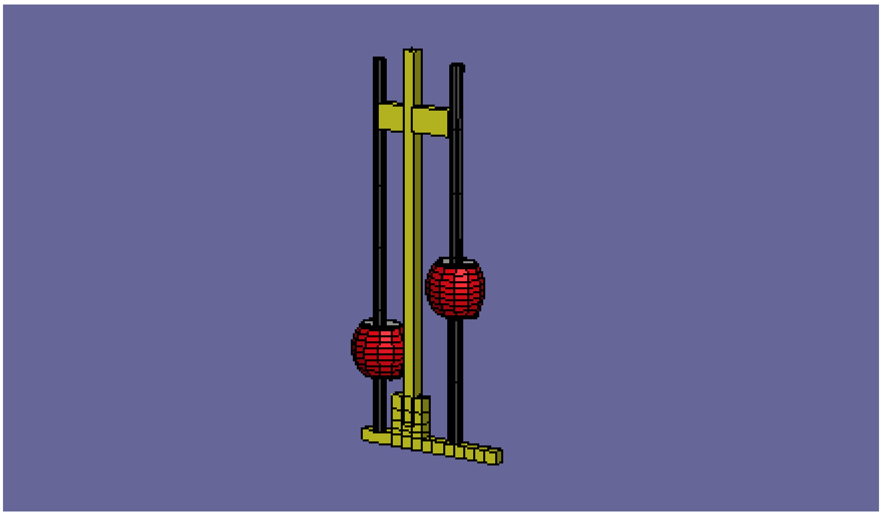
Conceptual Design of AFO (Top View)



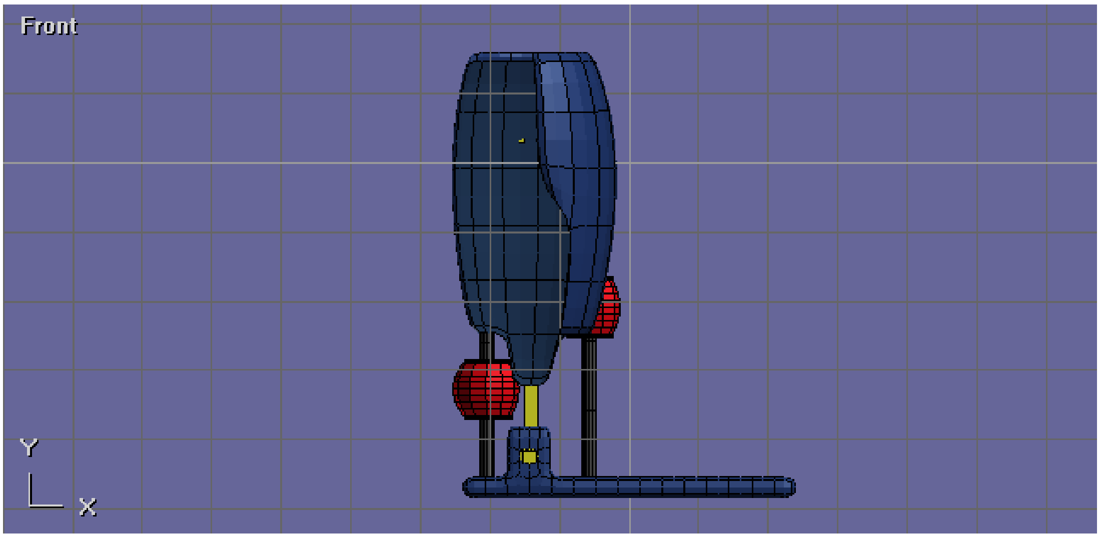
Conceptual Design of AFO (Left View)



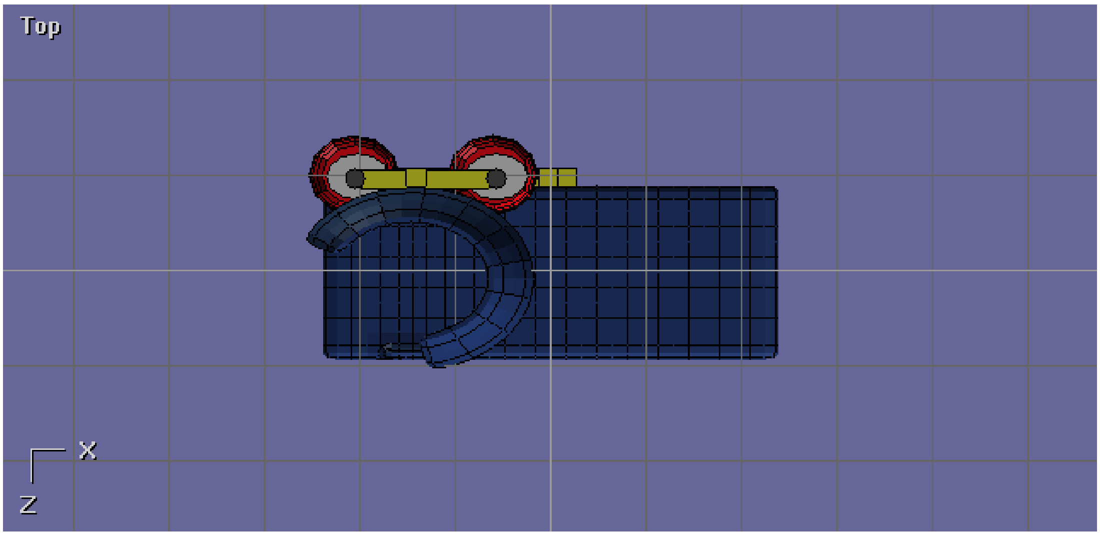
Conceptual Design of AFO (Perspective View)



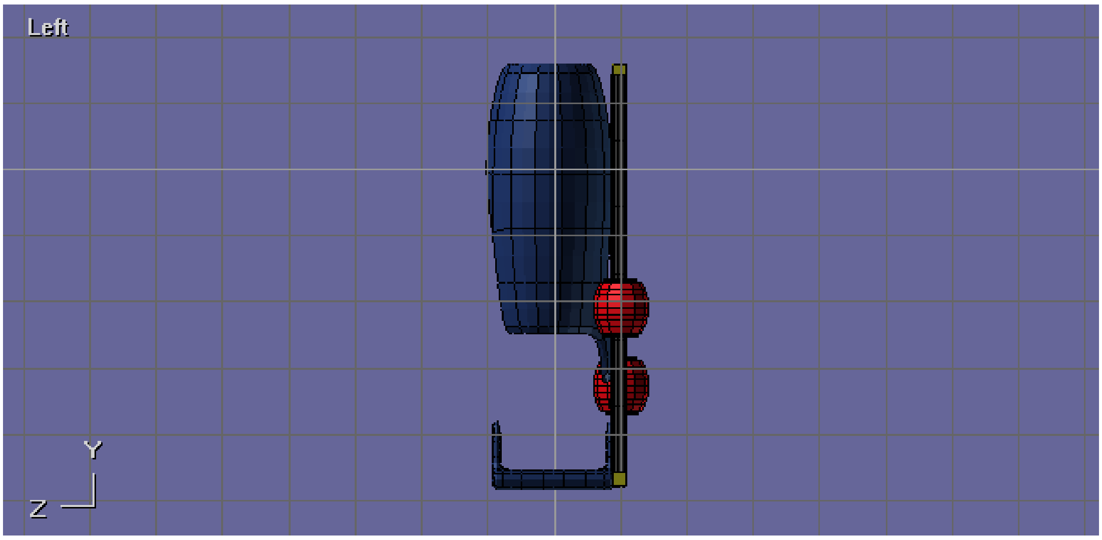
Conceptual Design of Frame for AFO (Red Colour Indicate the Pleated PAM)



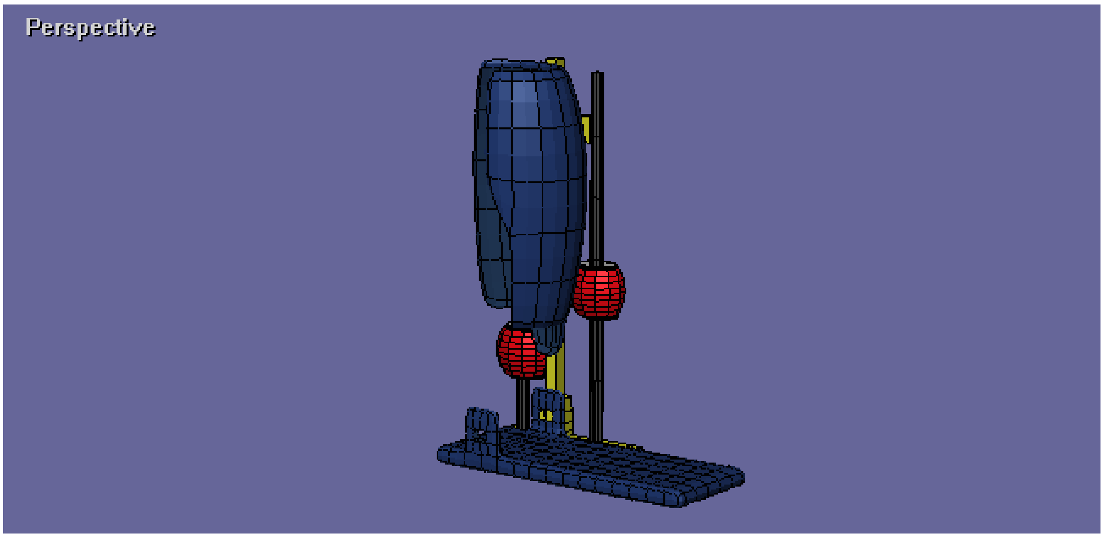
Conceptual Design of Prototype (Front View)



Conceptual Design of Prototype (Top View)

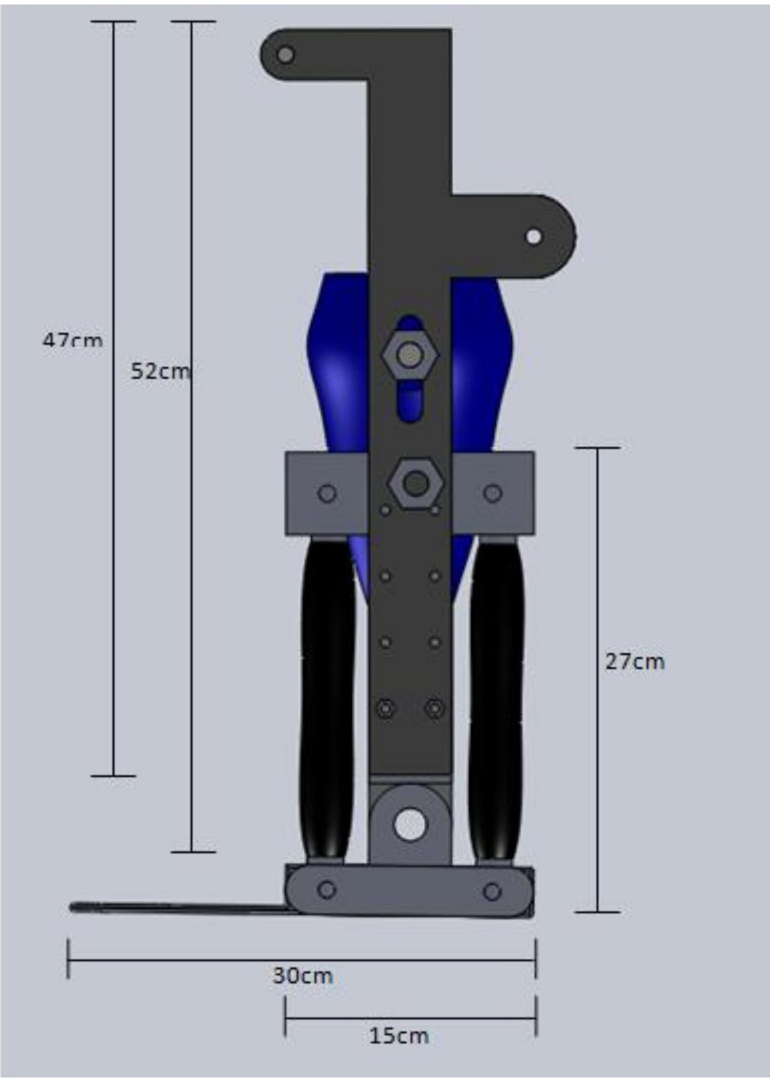


Conceptual Design of Prototype (Left View)

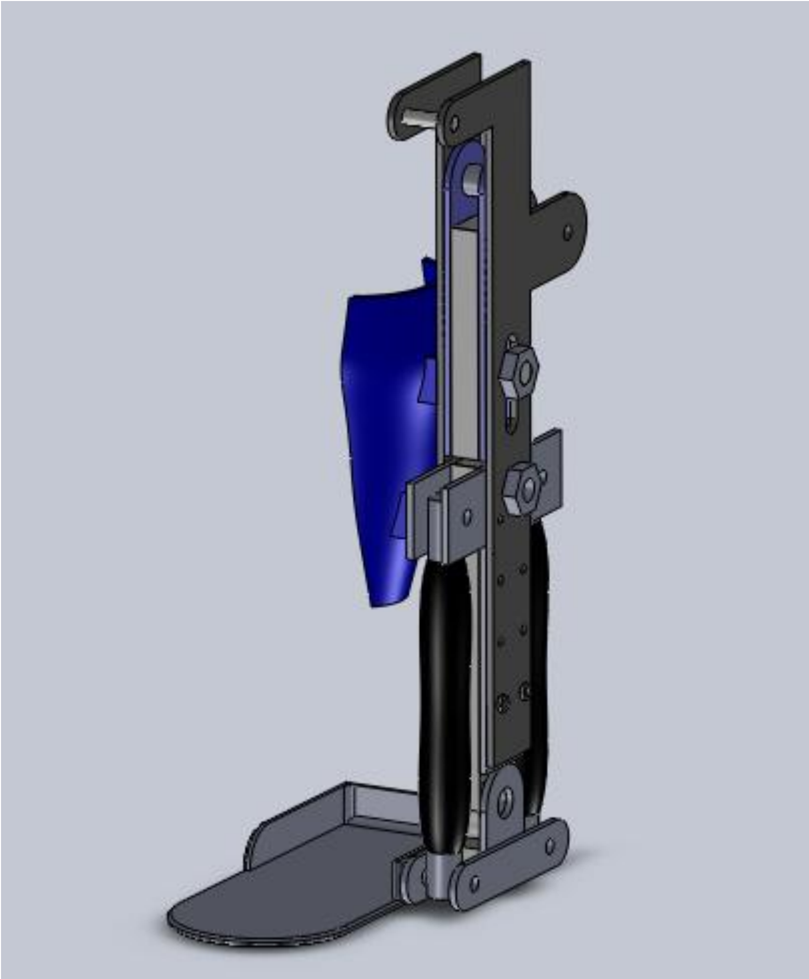


Conceptual Design of Prototype (Perspective View)

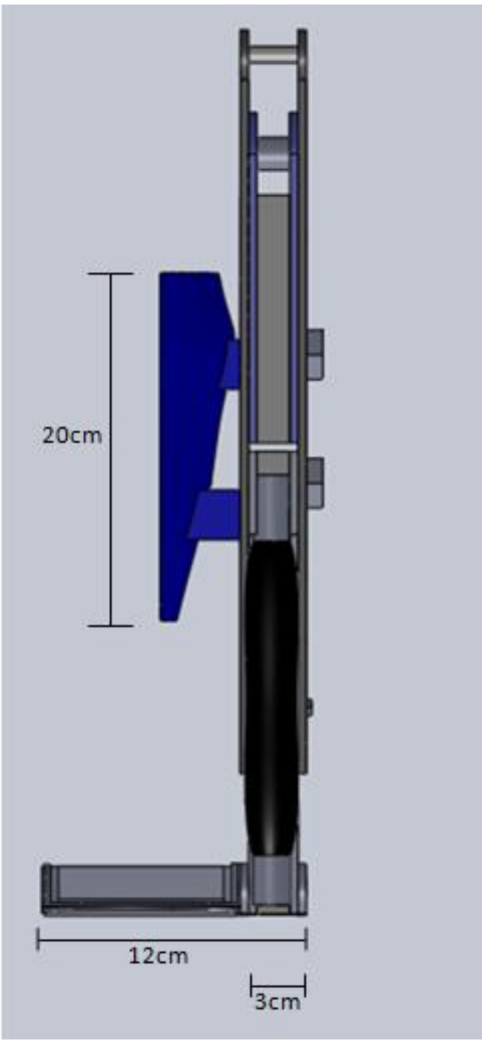
APPENDIX B: Conceptual Design Using SolidWorks



Dimensional Drawing (Front View)



Conceptual Design of prototype (Perspective View)



Dimensional Drawing (Right View)

APPENDIX C: Gantt Chart