ADAPTIVE NEUROFUZZY INFERENCE SYSTEM FOR AN ACTIVE TRANSFEMORAL PROSTHETIC LEG USING IN-SOCKET SENSORY SYSTEM

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ABSTRACT

Prosthetic leg is known as one of the solutions to help lower limb amputees to regain their ambulation ability. However, most of the current existing knee components still lack in the ability to provide active body propulsion, which in turn results in higher metabolic energy consumption required by the amputee in doing locomotion movement. This study was motivated by the idea of developing a microprocessor controlled prosthetic leg with user intention recognition for transfemoral amputees which is able to mimic a manner of ambulation close to that of healthy individuals. In particular, this study focused on developing the control input framework in order to derive the used intention, as read by the in-socket sensory system, as control input into the main controller. The first part of this work is the development of an Adaptive Neurofuzzy Inference System (ANFIS) - based algorithm to human gait phase recognition. The gait phase is necessary for cadence and torque control required by the knee joint mechanism. In the second part, the sensing method was adopted using the in-socket sensors embedded into the transfemoral socket of the prosthetic leg. A transfemoral amputee walked with in-socket sensorised prosthetic leg performed several gait cycles and the in-socket sensor signals were used as an input into the newly developed ANFIS system. Physical simulation of the controller presented a realistic simulation of the actuated knee joint in terms of knee mechanism, proving that the novel ANFIS system was validated with real amputee experimental signals. The fuzzy system successfully replicated human gait cycle by categorizing the cycle into seven gait phases. In conclusion, this study had established an ANFIS-based control input framework using in-socket sensory system based on amputee user's seven phases of gait cycle.

Keywords: Adaptive neurofuzzy inference system; prosthetic knee joint; controller; transfemoral leg; in-socket sensory system.

ABSTRAK

Kaki prostetik dianggap sebagai salah satu penyelesaian bagi membantu orang kurang upaya untuk mendapatkan semula kemampuan ambulasi mereka. Walau bagaimanapun, sebahagian besar komponen lutut yang sedia ada sekarang masih kurang kemampuan untuk memberikan pergerakan badan ke hadapan, yang seterusnya menyebabkan tenaga metabolik yang tinggi diperlukan oleh orang kurang upaya untuk melakukan pergerakan. Kajian ini adalah didorong daripada idea untuk membangunkan kaki prostetik mikroprosesor dengan menggunakan deria niat pengguna untuk orang kurang upaya atas lutut, di mana mereka dapat meniru cara-cara pergerakan seperti individu normal yang lain. Secara khususnya, kajian ini memberi tumpuan kepada pembangunan rangka kerja input kawalan untuk menghasilkan deria niat, seperti yang dibaca oleh sistem penderia soket sebagai input kawalan untuk alat pengawal utama. Bahagian pertama kajian ini adalah perkembangan algoritma berasaskan Sistem Inference Neophytes (ANFIS) untuk mengenali sistem fasa jalan manusia. Fasa jalan diperlukan untuk kawalan pergerakan dan daya turjah yang diperlukan oleh mekanisme sendi lutut. Di bahagian kedua, kaedah penderiaan telah digunakan dengan menggunakan soket sensor yang dilengkapi di dalam soket kaki prostetik atas lutut. Satu individu kurang upaya atas lutut telah berjalan dengan kaki prostetik yang ditampal dengan soket sensor dan telah melakukan beberapa pergerakan jalan dan isyarat soket sensor digunakan sebagai input ke dalam sistem ANFIS yang baru dibangunkan. Simulasi fizikal pengawal menunjukkan simulasi realistik yang digerakkan oleh sendi lutut dari segi mekanisme lutut, membuktikan bahawa sistem novel ANFIS telah disahkan dengan eksperimen bersama orang kurang upaya. Kesimpulannya, kajian ini telah menghasilkan satu rangka ANFIS melalui satu sistem deria soket berdasarkan gaya berjalan individu kurang upaya dan tujuh fasa kitarannya.

Keywords: Sistem Inference Neurofuzzy, lutut prostetik, alat kawalan, kaki palsu atas lutut, system deria soket.

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LIST OF SYMBOLS AND ABBREVIATIONS

ANFIS	:	Adaptive neurofuzzy inference system
FIS	:	Fuzzy inference system
TSK	:	Takage sugeno kang

CHAPTER 1: INTRODUCTION

The most common cause of a person undergoing amputation procedures is vascular disease followed by traumatic incidents. When a normal and healthy person lost a part of his limb, he or she is faced with unstable emotion, financial and lifestyle changes. The amputee would need a prosthetics device and medical services or consultation in order to adapt to the new lifestyle. Prosthetics is an artificial extension that replace the missing body part whether it is upper or lower body extremities or also can be both. The market price of prosthetics is quite high in developing countries including Malaysia because there are yet any prosthetics components that are locally manufactured yet. These components are imported from developed countries such as America, Taiwan and Germany. Furthermore, the cause of high cost in providing the amputee the prosthesis is because lack of professional prosthetist in Malaysia and most of the prosthetist are foreigners. The professional prosthetists are required to provide proper fitting, constructing and alignment in order for the amputee to use the device comfortably. Studies by World Health Organization indicated that the current shortage of technician supplies is by approximately 40,000 and it would take about 50 years to train 17,000 more skilled professionals.

1.1 Overview

Lower limb prosthetics should effectively restore the mobility function of an amputee. The development of lower limb prosthetics has brought better joint mechanism and structures. Prosthetics have been proved to improve lower limb mobility of amputees by offering wide variety of prosthesis and orthoses. However, the transfemoral amputees' gait still has noticeable abnormalities, and the prosthetic legs generally have limited abilities to perform active body propulsion, and this was due to the problem in walking cadence control. Self-confidence of the patients also depends highly on the implemented control system of the prosthesis provided. Despite the evolution in prosthetic technologies

(Aeyels, Petegem, Sloten, Perre, & Peeraer, 1995; Arieta, Katoh, Yokoi, & Wenwei4, 2006; Segal, Ava D,Orendurff, Michael S,Klute, Glenn K, McDowell, Martin L Pecoraro, Janice a Shofer, Jane Czerniecki, Joseph M., 2006; Zahedi, Sykes, Lang, Cullington, & Zahedi OBE, 2005; Zhong et al., 2014; Zlatnik, 1998), most of the prosthetic devices are still considered as passive devices and couldn't helping much.

Furthermore, in other locomotive functions such as climbing up the stairs and slopes, amputees need significant power to control the knee and ankle joints. Moreover, for an above knee amputees they tends to consume 50% more of metabolic energy compared to healthy person, in order for them to do their walking ambulation (Fisher & Gullickson, 1978; Huang et al., 1979; Inman, 1967). The additional power are needed in order for them to overcome it and it can be done by means of external energy sources that should be integrated in prosthetic components such as the microprocessor knee (Aeyels et al., 1995; Segal et al., 2006; Thiele, Westebbe, Bellmann, & Kraft, 2014). In order to compensate for the required robustness of the external power sources, there are a few computer-controlled prosthesis nowadays that can adjust the movement of the prosthetic knee joint to any speed and compensate any gait deviations and amputee's usage of metabolic energy. The most modern generation of controlled prosthesis (Herr & Wilkenfeld, 2003; Kusagur, Kodad, & Ram, 2012; Zahedi et al., 2005) with computer controlled phases of walking gait in various condition are considered to be an important step forward to achieved gait pattern that has closest proximity to natural walking.

1.2 Problem statement

Prosthetic leg has been known as one of the popular options that can assists the amputees to regain their moving abilities by wearing prosthesis as substitution of the missing leg. However, study proves that there are numbers of identified adaptations that have to be made by prosthetic users. One of them, is the lack of plantar flexion power or push off power by the prosthetic leg. Thus, more work is forced to be done by the hip joint to facilitate the missing forward propulsion affect while performing gait. Consequently, eccentric work at the hip joint is larger and joint power during concentric knee extension is smaller as compared to the normal gait.

Therefore, an intelligent and smart prosthetic leg that can assist the movement and facilitate the forward propulsion effect which can lessen the energy required by the amputees to move with prosthetic leg is needed. The assisting movement can be easily implemented by adding the motors or actuators to the knee and ankle joint which can perform the desired joint angle and joint torque. However, without knowing the intention of the user while using the prosthetic leg whether they want to walk, climb the stairs or stand up from sitting, there is no way the assisting method can perform efficiently. Hence, a smart prosthetic leg should have a good sensory system that can helps the system to understand the intention of the user, so that respective assisting movement by the actuators can be performed accordingly.

1.3 Objective

1. To design and develop a signal classification criterion using Adaptive neuro based fuzzy inference system (ANFIS) for an active prosthetic knee prosthesis using seven phases of gait state.
2. To stimulate and test the ANFIS based on the active knee prosthesis motor feedback using seven phases of gait cycle.
3. To validate the control framework of the develop ANFIS system using experimental signals of the in-socket sensory system by analyzing different kinetic and kinematics parameters in different activities of transfemoral amputees.

1.4 Potential impact of the research

An active prosthetic knee with good functionality for a fraction of the cost of a standard prosthesis is what developing countries and war-torn nations desperately need. More significant will be possible social impact of the compliant prosthetic knee. The specific control input framework advancements achieved through this work may be possible to use in other non-prosthetic rehabilitation applications like muscles stimulation for paralyzed people. The wireless communication methodology between sensing system and the controller, which is achieved through this research, can be also used in other biomechanics applications. This thesis may also direct to continued development to the ankle as well as to research in other compliant prosthetics. The technology used in this study may be used in developing a low-cost active ankle joint in future.

1.5 Thesis organization

This thesis project is to design and developed an Adaptive neuro fuzzy inference system (ANFIS) for an active transfemoral prosthetic leg using in-socket sensory system without deteriorate its functionality.

Chapter 2 covers with literature reviews of the past research, begins with a brief overview on gait phase and knee anatomy, and end with prosthetic knee joint design in the market.

Chapter 3 covers with the proposed control framework of an active prosthetic leg; different fuzzy systems are also investigated in this chapter to select a suitable knowledge-based system. In addition, the proposed ANFIS control framework would be tested with the in-socket sensory system via experimental with the amputee.

Chapter 4 covers with the discussion regarding the results obtain in this research.

Finally, **Chapter 5** provides the conclusions of the thesis and accounts for recommendations for future academic and commercial endeavor.

CHAPTER 2: LITERATURE REVIEW

Nowadays, there is wide variety of design incorporated with high technologies features in prosthetic knee joint available in the market. The variety design is made based on different goals that they want to achieve in order to meet the users demand. Some of the prosthesis are made to give the best cosmetically appearance of physiological limbs while others are made to provide the good functionality in order to do the daily activities living. Some user prefers cosmetic appearance to look same as near as possible to normal limbs as priority rather than functionality because of point of view from society around them about prosthesis. In this era, the technology of lower limb prosthetics is widely researched and become more variety in terms of design and functionality to provide amputees the benefits of advances in prosthetics to overcome the current technological limitations of amputee total restoration. A microprocessors-controlled technology able to learn and produce the pattern of the human anatomical control system, achieve beyond of the abilities of typical mechanical-based prosthetics. The system needs to incorporate sensor input, processing, output actuation, and feedback input features. The latest advance technologies in computer processing, materials, and actuation strategies have widely used resulting more prosthetics design to reshape what an amputee's experience is like after the loss of limb. In this study, I will focus on developing an algorithm for an active prosthetic knee joint in which there are design to give good performance on mimic the human anatomical control system and durability with a low cost. The good prosthetic knee joint is focused on reducing the production cost and maximize the performance simultaneously depends on its control input and its sensory system and follows the stride of physiological gait of normal human walking.

2.1 Anatomy defined

Human body are divided into two parts which are upper limb and lower limb of human body. While, on lower limb of human body, there are four main bone existing between the hip and ankle joint which are femur, patella, tibia and fibula. These four main bones contribute the most in human gait ambulation. The femur bone is the longest and strongest bone existing in lower limb in which it extends from the pelvis and straight to the knee. Meanwhile, tibia and fibula are two long bones in human leg existing between the knee and ankle joint. The upper end of tibia will join with the lower end femur to form the knee joint, which is the most crucial joint on lower limb of human body.

On the other hand, femur has two lower rounded ends or called as condyles, the one towards the center of human body is called medial condyle whereas the one to the outside is called lateral condyle. Above the condyles on both sides are epicondyles which work as sites for muscle and ligament attachment. The cruciate ligaments attach to the space between the two condyles is called intercondylar fossa. Cruciate ligaments are the most important ligaments in the knee joint and they serve to stabilize it and guide its motion. The patella which also called as knee cape existing to protect the joint and increases the quadriceps lever arm thus allowing the quadriceps to apply force to the tibia more effectively during extension. This triangular-shaped bone is not connected to femur and tibia directly. The patella is connected to the femur by being contained within the patella tendon that connects the quadriceps muscle to the tibia. Fibula has no contact with the knee and attaches to the tibia by ligaments below the tibia bearing surfaces of the knee (Dunn, 1987; Extremities, Lateral, & Extremities, 2005; Gray, 1918; Mcgrouther, 1991; Stevens, 2006; Yąsar İşcan, 1993).

2.1.1 Movement of the knee

Figure shows the three axes and planes of rotation of the biological knee joints. The anatomical planes allow for position or orientation representation of the knee in any of its original planes. The line connecting medial and lateral femoral condyles defines flexion-extension motion, $\emptyset$. The line along the tibia determines the axis of rotation for the internal-external angle, Ψ. The perpendicular axis to the other two axes defines as the abduction-adduction angle, θ.

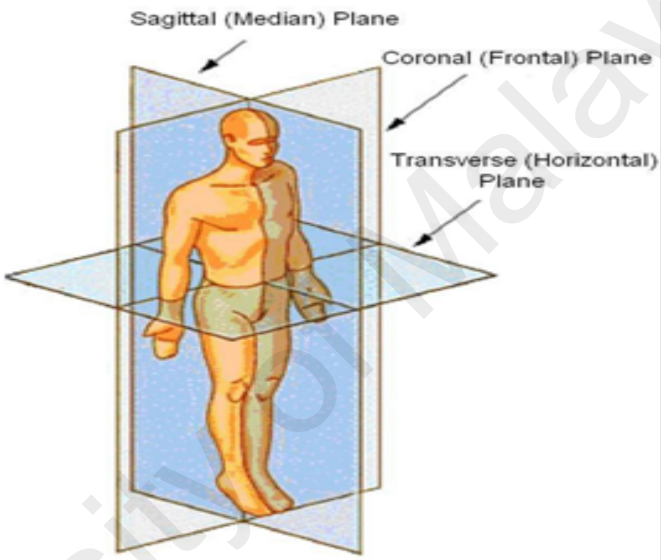


Figure 2.1 : Anatomical planes of human body

I. Sagittal (median) plane

An upright plane passing from front to back; separate the body into right and left halves.

II. Coronal (frontal) plane

A perpendicular plane running from side to side; split the body into anterior and posterior parts.

III. Transverse (horizontal) plane

A flat plane; divides the body into upper and lower portion of the body.

2.2 Human gait cycle

Gait is an aspect of kinesiology that involves the study of walking or other type of ambulation. The walking pattern of a person starting from the aspect of their kinematics of the body and body segment as well as the force acting on it (D. A. Winter, 1991; Whittle, 2002). Walking is the most convenient way to travel short distance. Free joint mobility of the body and appropriate muscle force can increase walking efficiency. As the body moves forward, one limb typically provides support while the other limb is advanced in preparation for its role as the support limb. Gait is an important sign of health and disease. A person walking with toes turned out at right angles may be suffering from flat fleet, or the gait may be due to stiffness. That is the reason why gait analysis plays an important role in understanding biomechanics of human body. Gait analysis is the study of the components of a stride of a person walking or running.

The act of walking has two basic requisition which are the periodic movement of each foot from one position of support to the next and must have sufficient ground reaction forces applied through the feet in order to support the body during walking. This periodic leg movement is the essence of the cyclic nature of human gait (Figure 2.2). Conventionally, a cycle begins when one of the feet makes contact with the ground. In normal gait, each gait cycle is divided into two periods which are stance and swing phase. Throughout a normal walking gait, repetitive events or cycle occur starting from heel strike and end at toe off. Since the events occur in a similar sequence and are independent of time, the gait cycle can be described in terms of percentage, rather than time. Thus, it allows normalization of the data for multiple subjects.

The initial foot strike occurs at 0% and the second foot strike occurs at 100%. The opposite leg undergoes the same events but one of the phase by 180 degree, with the opposite foot strike representing the 50% mark (Rose, J.G.Gamble, 1994). Each stride represents one gait cycle and is divided into two periods; stance and swing phase (Figure 2.3). Stance phase started when foot is in contact with the ground and constitutes 62% of the gait cycle. The remaining 38% of the gait cycle constitutes the swing period that is initiated as the toe leaves the ground. The stance phase is divided into four phase which are initial double support, midstance, terminal stance and second double support.

The initial limb support is characterized by a very rapid weight acceptance onto the forward limb with shock absorption and slowing of the body's forward momentum. Midstance and terminal stance are involved in the task of single limb support when the weight of the body is fully support by the reference limb. The second double support which is called preswing, prepares the limb to swing, transfer of body weight for the reference limb to the opposite limb takes place in this stage. Meanwhile, the swing period can be divided into three phase starts with the toe off and ends with the foot clearance when the swinging foot is opposite the stance foot. Midswing continues from the end point of the initial swing and continues until the swinging limb is in front of the body and the tibia is vertical. In the terminal swing, the limb is decelerated and finally strikes the ground for the second time. Limb advancement is performed during the preswing phase and throughout the entire swing period.

In traditional nomenclature, the stance phase events starting from the heel strike, move to foot flat, then to midstance, then to heel off and end with toe off. Heel strike is the one that initiates the gait cycle and represents the point at which the body's center of gravity at its lowest position. Meanwhile, foot flat is the time when the plantar surfaces of the foot touches the ground. Midstance occurs when the swinging (contralateral) foot passes the stance foot and the body's center of gravity is at its highest position possible. On the other hand, heel off start to occur as the heel loses contact with the ground and push off is initiated via the triceps surae muscles, which planta flex the ankle. Lastly, toe off starts when it terminated the stance phase as the foot leaves the ground. In swing phase, there are three events occur during the phase starting from acceleration, the events begin as soon as the foot leaves the ground and activates the hip flexor muscles to accelerate the leg forward. Then, the foot start to do midswing in which it occurs when the foot passes directly beneath the body, coincidental with the midstance for the other foot. Lastly, swing phase end at deceleration events in which starts when the action of the muscles as they slow the leg and stabilize the foot in preparation for the next heel strike.

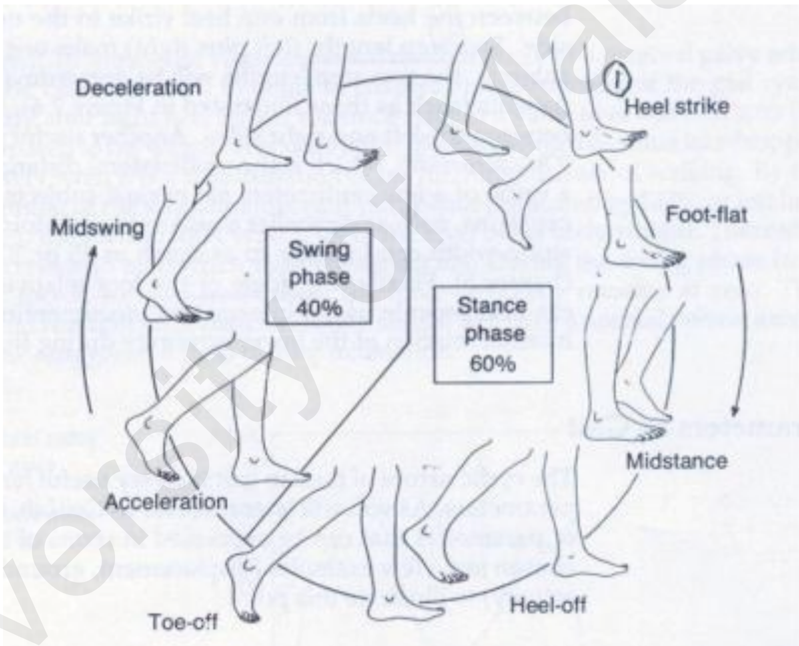


Figure 2.2 : Illustration of human walking gait analysis

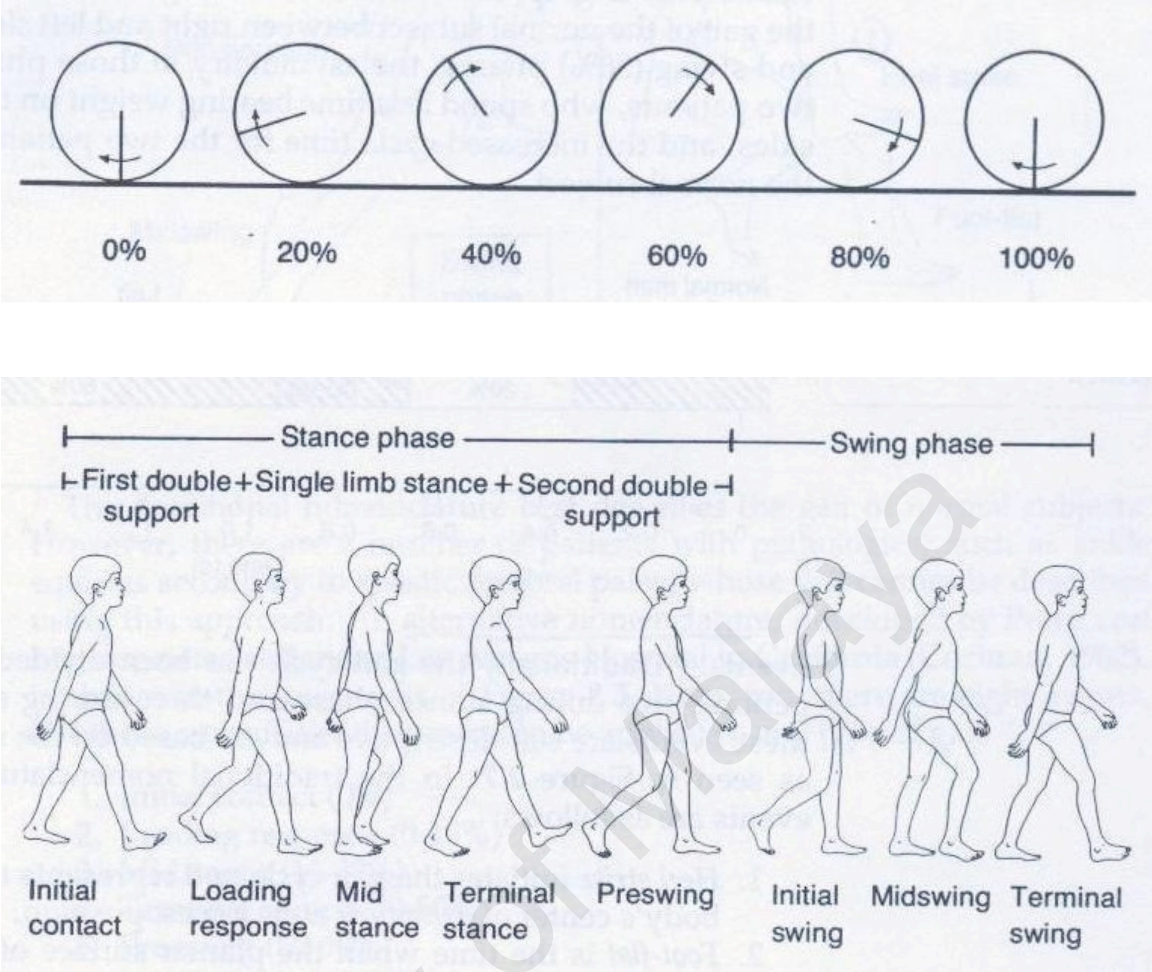


Figure 2.3 : Percentage of gait cycle together with their gait state

Note that through the nomenclature refers to the right side of the body, the same terminology would be applied on the left side. For a normal person, it is a half cycle behind or ahead off the right side. Therefore, the first double support for the left side and vice versa. In normal gait, there is symmetry between the left and right sides, but in pathological gait an asymmetrical pattern very often exists (Figure 2.4).

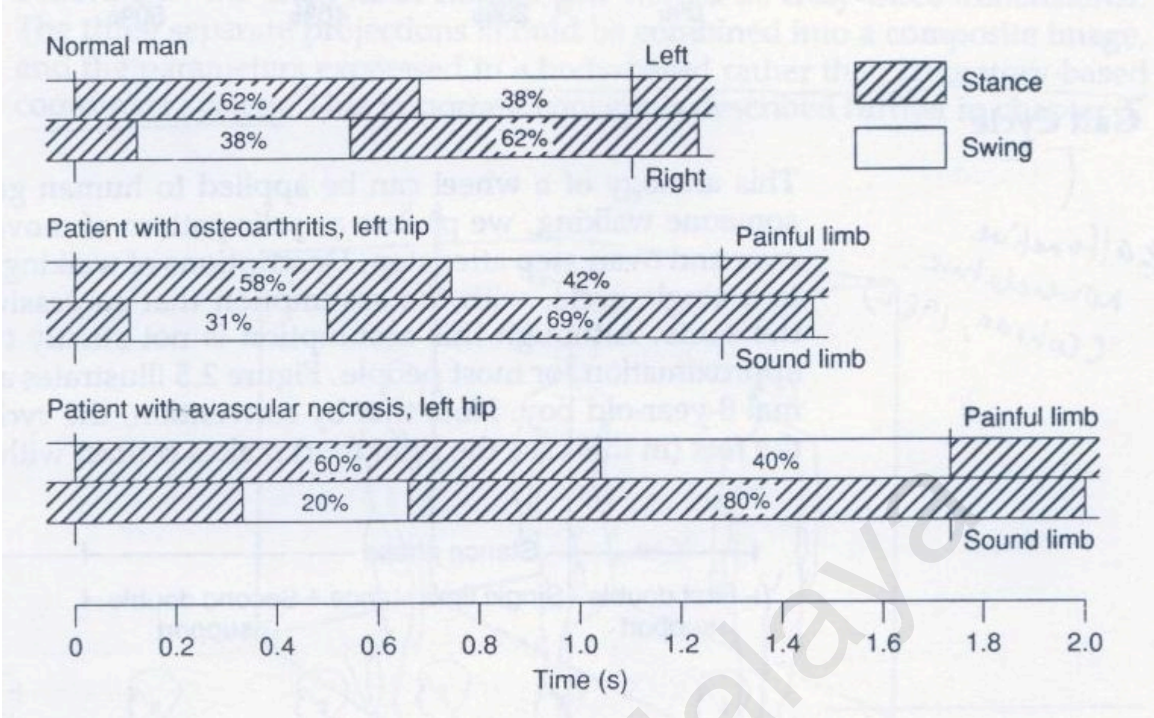


Figure 2.4 : Percentage of gait cycle for different people according to their health condition

2.3 Human compatibility

The active prosthetic leg device is non-invasive; hence biomaterial concern does not exist. However, the prosthesis must be able to withstand rigorous physical demands while also being light enough and durable for prolonged use. However, due to maintaining lower costs along with providing these necessary traits, the materials required a reasonable compromise.

2.4 Knee mechanics

A natural person walking gait is different and complex for every person depends on their anatomy structures and age. Thus, the current technology engineered devices cannot exactly replicate the true human ambulation and locomotion perfectly. Based on Fujie et Al (H. Fujie, G. Livesay, 1996), the human knee is comprised of 6 degrees of motion (DOF). The velocity of walking gait for a male adult transfemoral amputee with 70kg is near steady rate to 1.07 to 1.26 m/s according to M.L. Van der Linden, 1996. However, when the subject walking slightly faster gait, the velocity come nears to steady rate of 1.48 to 1.9 m/s. The subject takes about 1.58 to 1.75 stride length when walking. The human knee joint can be calculated by using Gruebler's Mobility Equation with Kutzbach's modification for planar mechanisms (Gogu, 2005). The overall number of degrees of freedom of the system can be calculated using the following equation;

$$\boldsymbol{M = 3(n-1) - 2f_1 - f_2} \qquad 2\text{-}1$$

Where M represents the degrees of freedom for the overall system, n, the total number of fixed link segments, f_1, the joints with one degree of freedom (DOF) and f_2, the joints with two degrees of freedom.

2.5 Overview on prosthetic knee joint

The current technology of prosthetics help amputees to do a lot of activities from low level to high level environmental barriers as the prosthetics provide good functionality. However, the high end prosthetic technology still have a significant gap to mimics the true human body mechanism, proving that there is a need for creative innovation and researches. The mechanical-based prosthetic system still cannot be compared with human natural movements of the limbs in all environment because lack of smooth and efficiency mechanism. A human body has brain as high end controller and processor to adapt and order proper instruction to the limbs as actuator according to the environment condition. Therefore, a computer controlled processor or artificial intelligent prosthetic is required in order to follow the natural human gait in various condition and all environment barriers (Jay Martin, BS, CP, LP Andrew Pollack, BS Jessica Hettinger, BS, Microprocessor Lower Limb Prosthetics: Review of Current State of Art)

Powered knee joint prosthesis has been introduced to restore gait and to overcome the issue of robustness with most of the passive devices, such as the well-known prosthetic mechanism agonist-antagonist active knee prosthesis (Martinez-Villalpando & Herr, 2009; Martinez-Villalpando, Weber, Elliott, & Herr, 2008). The aim of this study is to design a biomimetic active knee prosthesis that can follow true knee biomechanics level ground walking and utilizes series elasticity to minimize net energy consumption. The study shows that the passive knee and dynamically damped knees are by their nature capable only of negative mechanical power and therefore cannot replicate the generative phases of the natural knees (Johansson, Sherrill, Riley, Bonato, & Herr, 2005). Unfortunately, the design was compromised by inadequate reliability and complexity in daily fitting and maintenance requirements. The achieved gait is marginally improved compared with that of individuals who use conventional ones. The deterioration of the required locomotion of the transfemoral amputees require high metabolic energy consumption compared with healthy subjects.

When the required locomotion of transfemoral amputees is deteriorated, resulting to asymmetric walking gait and need more metabolic energy consumption relative to healthy subjects. However, a powered prosthesis can manage these problems significantly but having difficulties to overcome the weight, energy efficiency and range of motion. These factors need to be solved in this research. A good knee prosthetic is able to replicate the true anatomical knee range of motion and the ranges of torque and stiffness during normal human walking. In term of design, the knee prosthetic should be smaller in size and lighter from the loss of limb. The table below shows the reasonable specification and design criteria for a prosthetic to mimic a true anatomical limb;

Table 2-1: Design criteria for prosthetic knee joint

Length	330 mm
Width	70 mm
Weight	3 Kg
Flexion Angle range	0-120 deg
Output torque	100 Nm

To overcome the inadequacies of existing knee prosthetic devices, they propose an agonist-antagonist active knee prosthesis with an architecture inspired by the muscle anatomy of the natural human knee joint. In particular, this architecture (figure 3) allows for independent engagement of flexion and extension series springs (k_F and k_E) so that joint position and stiffness are independently controllable. This allows the controller to utilize the passive dynamics of the system and store absorbed energy in the springs for later use, thereby increasing efficiency (Martinez-Villalpando et al., 2008).

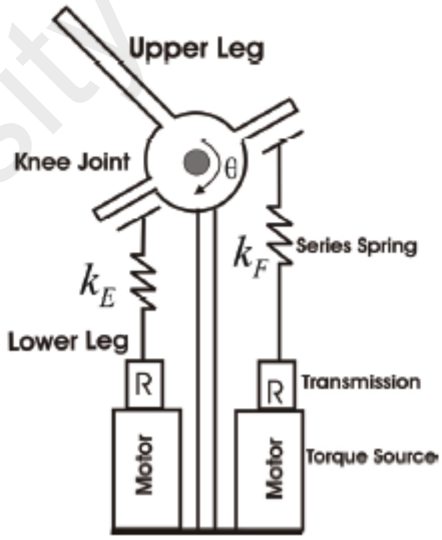


Figure 2.5 : Simplified mechanical architecture of the agonist-antagonist active knee prosthesis (Ernesto C.Martinez-Villalpando, 2008)

The actuation of the AAAKP is comprised of a pair of series elastic actuators (SEA), each of which consists of a torque source (active element) and a series spring, connected via a transmission. Two opposing SEAs are used to emulate the elasticity and damping characteristics of antagonistic muscle actuation. In level-ground walking, this design permits the use of a quasi-passive control scheme. Active elements are used primarily to independently control the engagement of each tendon-like spring, thereby controlling the transformation of potential energy into kinetic energy.

Since the ideal behavior of the knee on level ground is net dissipative, such a quasi-passive approach can in theory be extremely efficient. They hypothesize that this device can produce an adequate and appropriate positive power output at the knee joint, thereby reducing the network produced at the hip and, consequently, the metabolic cost of level ground walking (Martinez-Villalpando et al., 2008). Figure 4 shows the mechanical design of the AAAKP;

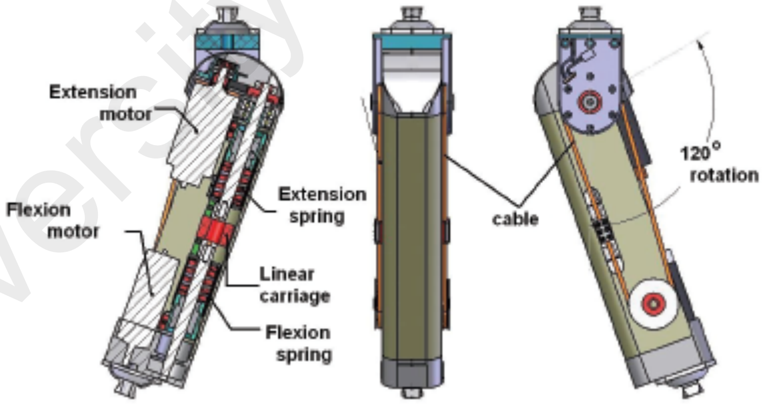


Figure 2.6 : Mechanical design of the AAAKP (Ernesto C.Martinez-Villalpando, 2008)

Other than that, there is one study that utilizes EMG as driven microcontroller based prosthetic leg (Islam, Haque, Amin, & Rabbani, 2011). The study involves redesigning of the motor and the gear system and that of the electronic circuitry for processing the EMG signals extracted from thigh muscles, interfacing the output to the microcontroller, rotating the motor in two directions thereby accomplishing the movement of the knee joint. The motor, geared down, is mounted horizontally and a pulley system drives the artificial knee joint.

Other than that, there is one study that shows the comparison of variable-damping and mechanically passive prosthetic knee devices (Johansson et al., 2005). The purpose of the study is to verify whether variable-damping knee prosthesis offers some improvements over mechanically passive prosthesis to transfemoral amputees. The results indicate that variable-damping knee prosthesis offer advantages over mechanically passive designs for unilateral transfemoral amputees walking at self-selected ambulatory speeds. Moreover, the results also further suggest that a magneto rheological-based system may have advantages over hydraulic-based designs. Here are some of the advantages of variable-damping which are decreases fatigue experiences during ambulation, are easier to maneuver and allow for smoother movement than mechanically passive knee (Kirker, Keymer, Talbot, & Lachman, 1996).

Modeling and simulation of self-tuning rapid-releasing absorber for above-knee prosthesis is one of the researches that been done to help transfemoral amputees (Mao, 2009). This study proposes a novel shock absorber design for an above-knee prosthesis, which provides automatic smooth tuning of the damping coefficient and rapid rebounds after impact loads. This research proposes an innovative self-tuning damper of a shock absorber for above-knee prostheses equipped in the tibia section, as shown in Figure. 5;

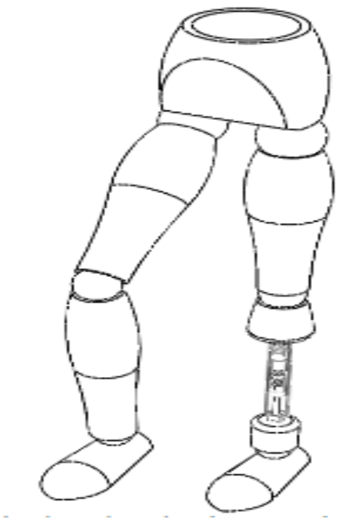


Figure 2.7 : The proposed shock absorber in the prosthetic tibia section (Y. C. Mao, 2009)

The fundamental idea of the self-tuning mechanism is to set up an orifice with a variable sectional area in the fluid passages inside the damper and pressure limiting check valves. This damper automatically tunes the damping coefficient by altering the oil flow rate through the passages according to the input force or velocity. It automatically locks out the shock absorber when required, specifically during acceleration when the amputee is during normal propelling. This mechanism automatically lowers the damping coefficient as well when the shock absorber encounters impact loads over heel strikes.

Furthermore, the check valves are rapidly opened when the patient is falling onto the ground or bouncing, producing the rapid contraction and rebound functionalities. This design achieves the automatic tuning by a pressure-sensitive plunger valve system and check valves without any electronic devices. Figure 6 shows the overall sectional view of the proposed design while figure 7 shows fluid circuit inside the damping sector;

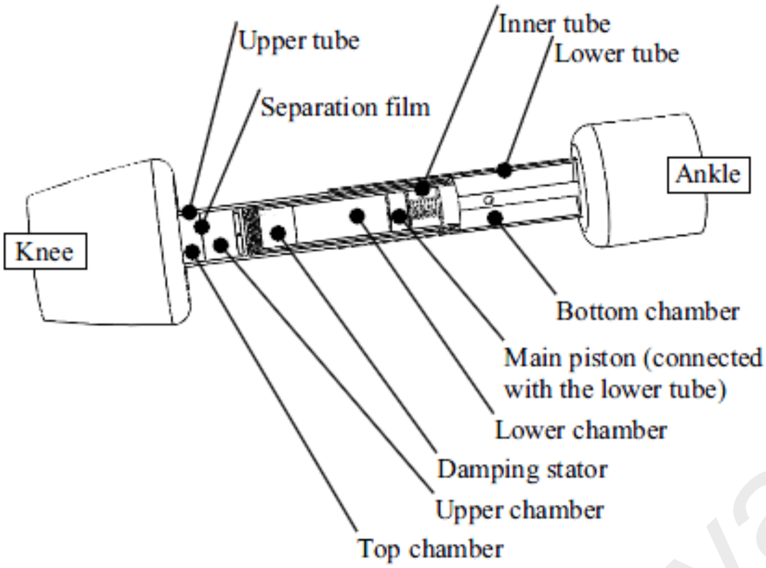


Figure 2.8 : The overall sectional view of the proposed design

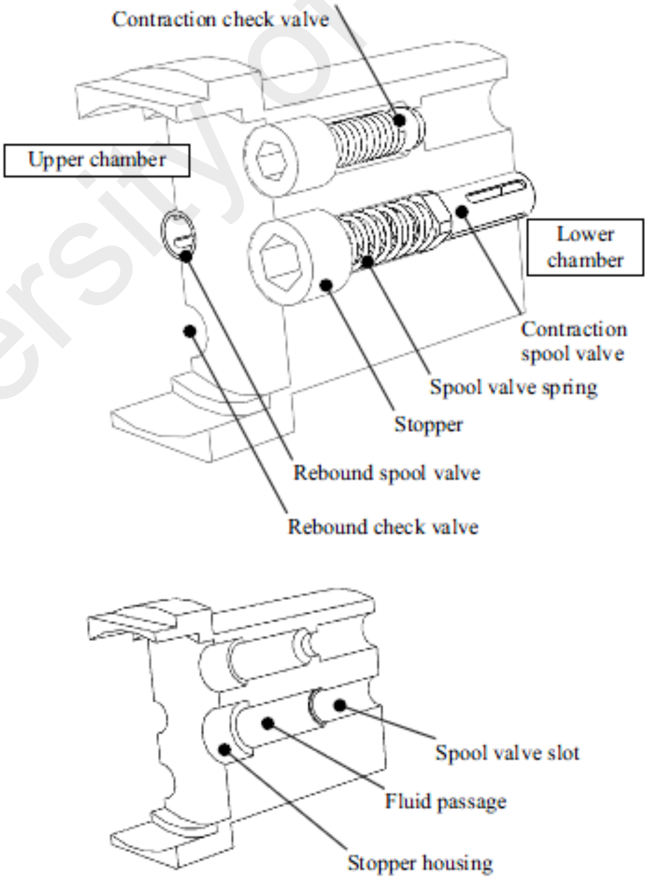


Figure 2.9 : Fluid circuits inside the damping sector

The most modern generation of controlled prosthesis with computer-controlled phases of walking gait under various conditions is considered an important step to achieve the gait pattern with the closest proximity to natural walking. A microprocessor-controlled technology can learn and produce the pattern of a human anatomical control system and achieve abilities beyond those of a typical mechanical-based prosthetics. This system should incorporate sensor input, processing, output actuation, and feedback input features. Although current prosthetics technology can help amputees perform many activities in various environmental barrier levels, high-end prosthetic technology has a remarkable gap to mimic the corresponding mechanism of a true human body, thereby confirming the need for creative innovation and research in this field.

Jasni et al. confirmed a reliable technique to extract a user's movement signal using a piezoelectric material as in-socket sensory systems in a transfemoral prosthetic leg. This study reported the efficacy of using a piezoelectric sensor as an alternative input to control actuators and its feedback system in the active prosthetic knee based on the force profile of muscles and the corresponding ground reaction force while performing lower limb movements. In addition, El-Sayed et al. proposed the use of piezoelectric transducers as a feedback signal source in controlling lower limb prosthesis. Both studies have verified that in-socket sensory systems can be used to characterize movements with remarkable sensitivity against the stump and provide an accurate signal of voltage arrays to be used as a control feedback system of an active prosthetic leg. Although both studies have presented the reliability of piezoelectric sensors to be used as an in-socket sensory system for user movement and intention detection, they have yet to verify that a signal can be used as an input to a feedback control system.

The importance of intention detection is to actively control actuators and to reduce the work load at the knee joint and the user's rate of metabolic energy. Therefore, this study focused on developing the control input framework to derive the user gait intention as read by the in-socket sensory system. This would be the control input into the main controller. In the first part of this study, an adaptive neurofuzzy inference system (ANFIS)-based algorithm that could derive human gait phase recognition was developed, considering that the gait phase is essential for the cadence and torque control required by the knee joint mechanism. In the second part, the sensing method was adopted using the in-socket sensors embedded into the transfemoral socket of the prosthetic leg. The ANFIS system also gathered the input signal from the in-socket sensorized prosthetic leg worn by a transfemoral amputee who performed several gait cycles. Thus, this study established an ANFIS-based control input framework by utilizing the in-socket sensory system based on an amputee user's seven phases of the gait cycle.

Meanwhile, Table 2.1 summarizing the classification and comparison of input method, control, and actuation for currently available lower-limb computer-controlled knees and feet that been adopter from Jay Martin (Martin, Pollock, & Hettinger, 2010).

Table 2-2 : Summary of various knee existing in current market

Device	**Input method**	**Sensor input**	**Control methods**	**Actuation method**	**Additional unique control features**
Knee Otto Bock (Duderstadt, Germany) C-Leg	CIC	Knee angle sensor, force-sensing strain gauge, time counter	Real-time control and variably resistive in stance and swing	Hydraulic valve actuator	Hydraulic valve actuators have inherent hydrodynamics that account for terrain and speed changes within a narrow range, and microprocessor-controlled valves broaden this range. Practitioner programmable settings
Otto Bock Compact	CIC	Knee angle sensor, force-sensing strain gauge, time counter	Real-time control and variably resistive in stance, but not in swing	Hydraulic valve actuator	Controls for a limited gait cadence. Practitioner programmable settings

Ossur (Reykjavik, Iceland) Rheo	CIC	Angle sensor, force-sensing stain gauge, time counter	Real-time control and variably resistive in stance and swing	Magnetorheological actuator	Self-learning program. Practitioner programmable settings
Ossur Power Knee	CIC with using IEC sound foot inputs	Sound Limb gyrometers, pressure and load cells to measure motion, position, and velocity, Bluetooth transfer, angle sensor, time counter	Real-time control and variably powered in stance and swing	Electromechanical actuator	Replaces lost concentric muscle function to restore symmetry, balance, and power. Offers spatial orientation of the limb. Practitioner programmable settings
Freedom Innovations (Irvine, CA) Plié MPC	CIC	Extension sensor, force-sensing strain gauges, time counter, step counter	Microprocessor management of stance-swing transition, hydraulic control of resistance during swing	Hydraulic valve actuator	Practitioner programmable settings

Endolite (Centerville, OH) IP Plus	CIC	Angle sensor, ambulation speed sensor, time counter	Variable resistive actuation in swing, mechanical breaking weight-activated stance	Pneumatic piston	Practitioner programmable to specific activities, cushion valve, energy management
Endolite Smart IP	CIC	Angle sensor, ambulation speed sensor, time counter	Variable resistive actuation in swing, mechanical breaking weight-activated stance	Pneumatic piston	Practitioner programmable to specific activities, self-learning
Endolite Smart Adaptive	CIC	Angle sensor, force sensor, time counter	Variable resistive actuation in swing and stance	Swing–pneumatic piston, stance–hydraulic actuator	Practitioner programmable to specific activities, automatic modes for stairs, stance, stumble recovery, standing, accommodates ambulation speed and slopes
Daw Industries (San Diego, CA) SLK	CIC	Electromagnetic position, velocity, and angular momentum sensor, microcontroller time sensor collects signal in 3D grid space	Variable resistive actuation in swing	Pneumatic actuator	Self-learning program, records last 100 steps in 11 different ambulation speed categories, readjusts categories if user exceeds maximum speed, five-bar geometric design for enhanced smoothness of gait and stance stability

Feet Ossur Proprio	CIC	Angle sensor, accelerometer, time sensor	Variable position, swing phase actuation only for stance and swing modes	Electromechanical actuator	Sitting mode, with increased plantarflexion, and swing phase dorsiflexion mode. Does not alter resistance, but only angular position

In addition, Table 2.2 summarized the classification and comparison of the sensory input and control methods of the currently available lower-limb prosthetic knee joints existing in the market;

Table 2-3 : Summary of various knee joint with its sensor inputs and control methods

Device	***Sensor Input***	***Control methods***
OTTOBOCK knee (c-leg, genium)	Angle, force sensor, time counter	Microprocessor control
Ossur rheo knee	Angle, force, time counter	Real time control with self-learning programme
Ossur power knee	Gyrometer, load cells, angle, time counter	Real time control
Freedom innovation	Force, strain gauge, step counter, time counter	Microprocessor with hydraulic control
Endolite IP PLUS	Angle, ambulation speed, time counter	Mechanical breaking weight-activated stance control
Endolite Smart IP	Angle, speed, time counter	Mechanical breaking weight-activated stance control
Ossur proprio	Angle, accelerometer, time	Mechanical control with electromechanical actuator

2.6 Gait of transfemoral amputees using prosthetic knee

Mobility plays an important role in individual's quality of life and daily activities living. Unlike a person with disabilities, they lost some of their mobility to do their daily activities living depends on what kind of amputation they previously undergo. Moreover, their walking gait also would be affected and the severity of it depends on the quality of surgery had been done by the orthopedics. For an instance, how well the condition of the stump is, and how much the length of the remaining muscular structure and how well they are reattached. This is the reason why, gait plays an important role in amputees' rehabilitation to restore back their quality of life.

Prosthetic device implementation is one of the ways to help the amputees to gain back their ambulation (Buckley, De Asha, Johnson, & Beggs, 2013). Prosthetist need to be aware of amputee's gait in order to implement prosthetic device to them in order to prevent knee buckling from happening since it is the common gait deviation during walking (Torres & Esquenazi, 1991).

In order to overcome it, a 'free knee' need to remain in extension position for about 30% to 40% of stance phase to counter act this issue (Pezzin, Dillingham, & MacKenzie, 2000). This to ensure that the heel strike is prolonged enough to move the body forward over the prosthetic leg device during stance phase state. Meanwhile during this time, the hip extensor was used to stabilize the prosthetic leg device so that the amputees can have full weight bearing condition (Segal et al., 2006). At the same time, during swing phase, hip extensor and calf muscles on the residual limb will help to generate enough forces in order to bring the contralateral leg to swing forwards. Moreover, hip flexor of the residual limb also must generate the same amount of force in order to achieved the normal gait required (Perry, 1992; Whittle, 1996).

Despite all that, amputee's control and strength are reduced due to shortening length of thigh muscles, which also will reduce the force of muscle contraction. They are various kind of knee joint provided in current market and one of them are called fixed knee joint means the knee are unable to do flexion and extension motion. For amputees who wearing this kind of knee, they might face some difficulty in floor clearance during swing phase, due to lack of knee flexion and ankle dorsi flexion. To overcome this shortcoming, the amputee's need to use their hip muscles to elevate the hip in order to prevent foot dragging or called as hip hiking.

Thus, the overall energy expenditure for transfemoral amputees is higher than those who undergo below knee amputation since they still have knee joint to the ambulation and does not required external forces from the hip and trunk muscle to move the contralateral leg.

There is various treatment that can be used to overcome the shortcoming of transfemoral amputees and one of them is through rehabilitation technique. Not all of techniques provided is suitable for each amputee, therefore the program or technique applied must be reviewed regularly to ensure it remains adequate. All the techniques and program provided are depends on amputee's previous level of activity and also their current health overall condition in order to restore back their ambulation mobility. In order to overcome it, there some precaution or circumstances need to be taken such as during assessments of the amputees, prosthetist or doctor must be aware of biomechanics gait of normal human gait and also how the gait of amputee would affect after the amputation. For an instance, the amputees might have experienced muscle weakness or tightening when trying to get used with the prosthesis provided. Moreover, some of them also might get lack of balance and fear during wearing the prosthesis and that is the reason why the prosthetist need to be aware of it, thus a successful rehabilitation programmed can be achieved (Fairburn et al., 2002).

CHAPTER 3: METHODOLOGY

The motivation of this study was to develop a new rule-based written language for an active prosthetic leg by using in-socket sensory system as its main input control. This chapter consist of 2 major parts; the first part describes the development of new method for human gait phase recognition for cadence control. The second part detailed out the experiments conducted to feed real data from in-socket sensory system into the developed ANFIS control input system.

3.1 Preliminaries

This chapter covers the principle of fuzzy set theory, operation and relations/composition that provides the background required to understand Fuzzy Inference System (FIS). First of all, fuzzy system was introduced by Zadeh (L. A. Zadeh, 1965). The concept he was introduced was ‘linguistic variables’ or called as fuzzy set which used as an extension of traditional crisp set theory.

3.1.1 Fuzzy rules

Fuzzy rules are set of IF-THEN rule statements that describe how the system works in which multiple of antecedents and consequent is depicted. For an instance, statement below describe that IF part of the rule showed the antecedent or called as premise. Meanwhile, the THEN part is described as consequent or conclusion of the statement. The connective operator such as AND or OR are used to form multiple antecedent of fuzzy rules, which are equivalent to fuzzy intersection and fuzzy union respectively. It must be pointed out that only one of connective operator is allowed to be used to connect two premises together.

The above conditional statements comprise fuzzy logic.

$$\boldsymbol{R_l : IF\ x_1\ is\ F_1^l \frac{AND}{OR} x_2\ is\ F_2^l \frac{AND}{OR}\ x_p\ is\ F_p^l\ THEN\ y\ is\ G^1} \qquad 3\text{-}1$$

Whereby, l is describing as rule index, F_j^l are fuzzy input sets in universe discourse U and G^l is fuzzy output set in universe discourse V.

3.1.2 Fuzzy inference engine

Fuzzy inference engine is used to combine the rules into mapping from fuzzy input sets to fuzzy output sets.

One membership function is allocated to each fuzzy IF-THEN statement such as IF A THEN B to measure the degree of truth of the implication relation between x and y. Minimum and product implication of t-norm are used specifically in this work to describe this membership function $\mu_{\mathrm{A\to B}}(x,y) = \mu_{\mathrm{A}}(x) \otimes \mu_{\mathrm{B}}(y)$. Therefore, the membership function of the earlier multi-antecedent rule can be stated as,

$$\boldsymbol{\mu_{\mathrm{R(l)}}(x,y) = \mu_{\mathrm{A\to B}}(x,y) = \left(\mu_{A_{F_1^l}}(x_1) * \mu_{A_{F_2^l}}(x_2) * \dots \dots * \mu_{A_{F_p^l}}(x_2)\right) \otimes \mu_{G^l}(y)} \qquad 3\text{-}2$$

Where * is ⊗ (s-norm) or ⊗ (s-norm) depends on the construction of the stated rule and $F_1^l * F_2^l * \dots \dots * F_p^l$, A and G^l, B then $R_l: A \to B$. Note that part 'x' is excluded from the above evaluation. This part is taken into account as fuzzification which will be discussed in the following. IF-THEN rule involve in multiple step which is the first one; evaluating the premise and then applied to the consequent. Evaluating the antecedent itself is divided into two sections which is converting the crisp input numerical data into fuzzy sets. Secondly, utilizing the fuzzy operators to evaluate the antecedent in terms of degree of membership to all input fuzzy sets.

3.1.3 Fuzzification

In any fuzzy system, one needs to deal with fuzzy sets to be able to implement different fuzzy techniques. Therefore, mapping the crisp numerical data, $x = \{x_1, x_n\} \in U$, into fuzzy sets, $A_i^* \in U$ is the first step. This step or process is called fuzzification and it can be achieved through fuzzifier process. There are two major types of fuzzifiers which are singleton, and non-singleton.

In singleton fuzzifier with support *x'*, the membership function that associates with A^* can be considered as crisp membership function, in other words,

$$\boldsymbol{\mu_{A*}(x) = \begin{cases} 1 & for\ x = x' \\ 0 & otherwise \end{cases}} \qquad 3\text{-}3$$

Where x' is basically the crisp input. In a non-singleton fuzzifier with support x', the membership function that associates with A^* is fuzzy membership function.

$$\boldsymbol{\mu_{A*}(x) = \begin{cases} 1 & for\ x = x' \\ 0 \le \mu < 1 & otherwise \end{cases}} \qquad 3\text{-}4$$

Thus, for *x''*, $\mu_{A*}(x') = 1$, and $\mu_{A*}(x)$ decreases from unity as moves away from *x'*. However, the singleton fuzzifier by itself may not always be adequate, especially when input data is contaminated by noise.

3.1.4 Fuzzy reasoning

At this time, to complete the evaluation of the antecedent of the fuzzy rule, $\mu_{A*}(x)$ must be put into action. Hence,

$$\begin{aligned}\boldsymbol{\mu_{B'}(y) = \mu_{A*\in R^{(l)}}(y)} \\ \boldsymbol{= sup_{x\in A_x}\lfloor \mu_{A_{x1}}(x) \otimes \mu_{A\to B}(x,y)\rfloor}\end{aligned} \qquad 3\text{-}5$$

Equation above is the correlation between the crisp input and output of one excited rule. In other words, $\mu_{B'}(y)$ is the output of fuzzy inference engine; however, the grade of membership functions of inputs, $\mu_{A*}(x)$ were defined in the fuzzifier, substituting two equation above gives;

$$\begin{aligned}\boldsymbol{\mu_{B'}(y)} &\boldsymbol{= \mu_{G^l}(y)} \\ &\boldsymbol{\otimes \left\{\left[sup_{x\in U}\mu_{A_{x1}}(x_1) \otimes \mu_{A_{F_1^l}}(x_1)\right]\right.} \\ &\boldsymbol{* \dots\dots.. \left[sup_{x\in U}\mu_{A_{xp}}(x_p)\right.} \\ &\boldsymbol{\left.\left.\otimes \mu_{A_{F_p^l}}(x_p)\right]\right\}}\end{aligned} \qquad 3\text{-}6$$

The final fuzzy set can be obtained by combining all the $B^l$from all the rules in the rule-base. One method to find the final fuzzy set is using fuzzy union or s-norm for all the attainedB^l. However, most of the time, the final fuzzy set is determined during defuzzification process.

3.1.5 Defuzzification

At the end of any fuzzy mechanism, we need to convert the fuzzy set that is the output of inference engine to a crisp point. The inverse fuzzy transformation process, which is used to map the fuzzy output variable to that of a crisp one, is call defuzzification. There are many defuzzifier in the literature, however there are no scientific bases for any of them (Mendel, 1995). The most important criteria to select any of offered defuzzifier in the literature is the expense of the computation. Two most popular defuzzifier are depicted below;

I. Centroid defuzzifier

$$\overline{\boldsymbol{y}} = \frac{\int_c \boldsymbol{y}\boldsymbol{\mu}_{\mathbf{B}}(\boldsymbol{y})\, \boldsymbol{dy}}{\int_c \boldsymbol{\mu}_{\mathbf{B}}(\boldsymbol{y})\, \boldsymbol{dy}} \tag{3-7}$$

Where c indicated the support of $\mu_{\mathrm{B}}(y)$

II. Height defuzzifier

$$\overline{\boldsymbol{y}} = \frac{\sum_{l=1}^{M} \overline{\boldsymbol{y}^{l}}\, \boldsymbol{\mu}_{B^l}\left(\overline{\boldsymbol{y}^{l}}\right)}{\sum_{l=1}^{M} \boldsymbol{\mu}_{B^l}\left(\overline{\boldsymbol{y}^{l}}\right)} \tag{3-8}$$

Where $\overline{y^l}$ is the center of gravity of the set B^l.

3.2 Development of an ANFIS for the control input framework of transfemoral prosthetic leg

Adaptive Neuro Based Fuzzy Inference System (ANFIS) was developed to control the actuator movement with respect to the feedback signal in which, it will develop human gait phase recognition necessary for cadence control. The framework was developed according to the rule-base written in natural language which enables the controller to adjust the motor torque automatically depends on the subject's gait mode and gait velocity. A systematic procedure of developing the ANFIS were presented in this study. Particular pattern of knee angle and knee moment was used as reference input to the ANFIS system to test and verified the behavior of the knee mechanism in many locomotion conditions. The efficiency of the ANFIS system was evaluated via simulation and experiment.

3.2.1 Lower limb inclination angle

Figure 3.1 illustrated the schematic diagram of human body in sagittal plane motion (thigh and shank) in which represent the hip and knee joints of normal human body. This human body segment is repeatable, fixed and it is reliable to be used as an input for the controllers. The angle of lower limb illustrated in the figure 1 are defined based on the study by Winter (D. A. Winter, 1991). Winter stated that, when the knee is in fully extend position, the degree of flexion will be zero. Thus, when $\theta_{thigh} > \theta_{leg}$ the knee is in flexion position meanwhile when $\theta_{leg} > \theta_{thigh}$ the knee is in extended position. This method study by Winter provides reliable technique for calculation human body measurement since the external disturbance can't affect the output signal. There are few assumptions that needs to be taken for this method;

I. The trunk must be in vertical position
II. The joint center of rotation is in fixed position
III. All the segments are rigid bodies

As for the inclination angle of femur and tibia, it can be determining through the equation stated below;

$$\boldsymbol{\theta_{femur}} = \boldsymbol{\theta_{hip}} + \boldsymbol{\theta_{trunk}} \quad 3\text{-}9$$

Where $\theta_{trunk} = 90^{\theta}$

$$\boldsymbol{\theta_{tibia}} = \boldsymbol{\theta_{thigh}} + \boldsymbol{\theta_{knee}} \quad 3\text{-}10$$

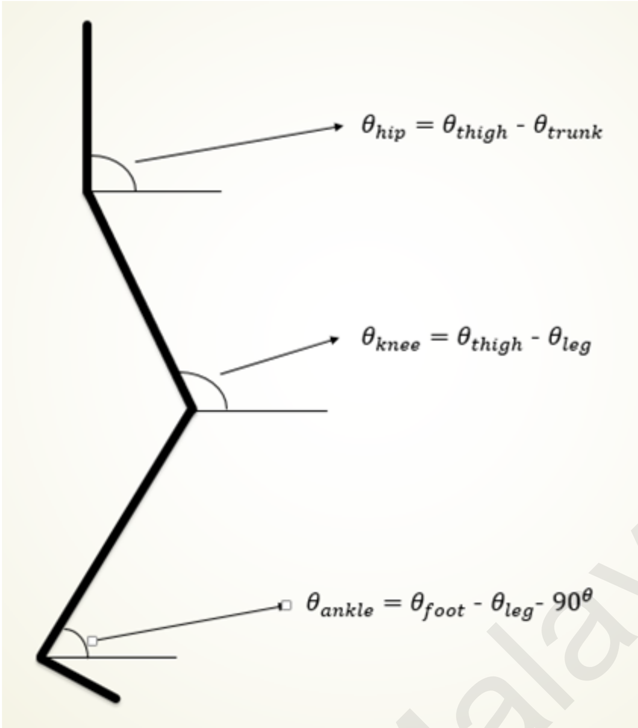


Figure 3.1 : The correlation between lower limb joint angles and segments inclination angles with the assumption that the joint centre of rotation is a fixed position point

3.2.2 Proposed Takage Sugeno Kang (TSK) FIS by using Least square method

A type-1 TSK FIS is a fuzzy system of network that was composed through five layers process. Input of the fuzzy system whereby the premise parameters are store will fed into the first layer. At this stated membership values of input are calculated and thus are sent to the next layer. Whereas, at the second layer, all the membership values are computes to firing level of rule based on the defined t-norm (Sugeno & Kang, 1988). At the third layer stage, the ratio of each rule's firing level to the sum of all rule's firing levels were calculated (L. Zadeh, 1965) and the outcome will be sent directly to the forth layer. Meanwhile at the forth layer, for each one of the rule will gives $f^1(x)$ $y^1(x)$ through accumulated consequent parameters by using least square optimisation method (Kosko, 1992). In order to use this method, some assumption need to be taken which are the shapes and parameters of all the input fuzzy sets are fixed ahead of time. Lastly, on the fifth layer, all the values are computes as a summation of all incoming inputs. In order to test the fuzzy system, the given training data are fed onto the network to achieve the desired input-output mapping. Therefore in this technique, although the antecedent parameters are fixed ahead of time, they will be tuned during the training process as will the consequent parameters. This iterative method continues until the sum of the squared errors over all the N training data becomes less than a predefined threshold.

Figure 3.2 showed the Takage Sugeno Kang FIS for the ANFIS system.

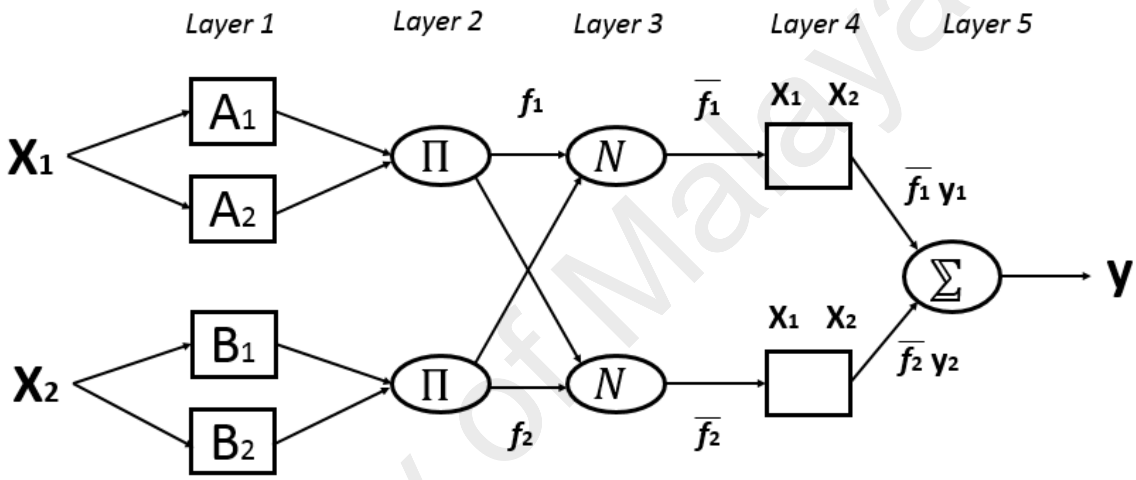


Figure 3.2 : TSK FIS structure in the form of ANFIS

Set of rules or fuzzy sets used in this system are based on seven phase of gait cycle, starting from the heel strike and end with toe off. The arrangement below is used for both input of inclination angle used in this system which are femur and tibia angle and all the fuzzy sets are fed onto the ANFIS system,

I. Phase 1 : Heel strike
II. Phase 2 : Midstance
III. Phase 3 : Terminal stance
IV. Phase 4 : Preswing
V. Phase 5 : Initial swing
VI. Phase 6 : Midswing

VII. Phase 7 : Toe off

3.2.3 ANFIS design and development

ANFIS is a rule-based written language that has been developed for gait control in which a modified knee trajectory from the residual limb was played back on the active prosthetic leg. Moreover, ANFIS also used to send correct value of torque to the motor controller so that the prosthetic leg can behave same like the contra-lateral leg. The magnitude of torque resulting from the ANFIS system were fed in with the angular position of the tibia and femur of the Winter data (Winter, 2009) to test the behavior of the ANFIS system. The advantage of these rule-base system is that it can eliminates the need of mathematical modelling of human body during implementation of the control process (Borjian, Lim, Khamesee, & Melek, 2008).

In this study, type-1 TSK was chose as the fuzzy system. In order to test the performance of the ANFIS, 70 data of femur and tibia angle from Winter database (Winter, 2009)was selected as the input or testing data for the fuzzy system. The purpose of using these data was to train the fuzzy system and see its performance regarding the how well the model can predict the corresponding data set output value. This fuzzy system was considered as two separated fuzzy system for each output which are knee angle (y^1) and knee torque (y^2) as shown in figure 3.3 and both outputs were trained individual despite, they have an identical input membership function.

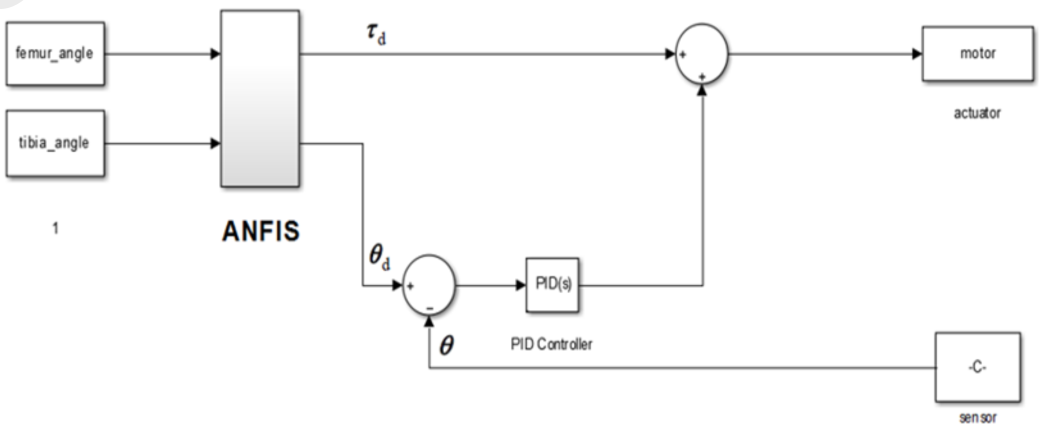


Figure 3.3 : The designed control diagram of prosthetic knee based on TSK FIS

They are few assumptions need to be taken for the fuzzy system to be works;

I. Data distribution in each cluster is normal and independent

II. The input membership function of the fuzzy system is in Gaussian type

III. 7 fuzzy sets were fed into the fuzzy system

In order to define the input parameters of the fuzzy system, the mean and standard deviation in each cluster of the data were calculated and the result depicted in Table 1 and Table 2. Meanwhile, the membership functions for the input femur angle (x^1) and tibia angle (x^2) were depicted in figure 3.4 and figure 3.5.

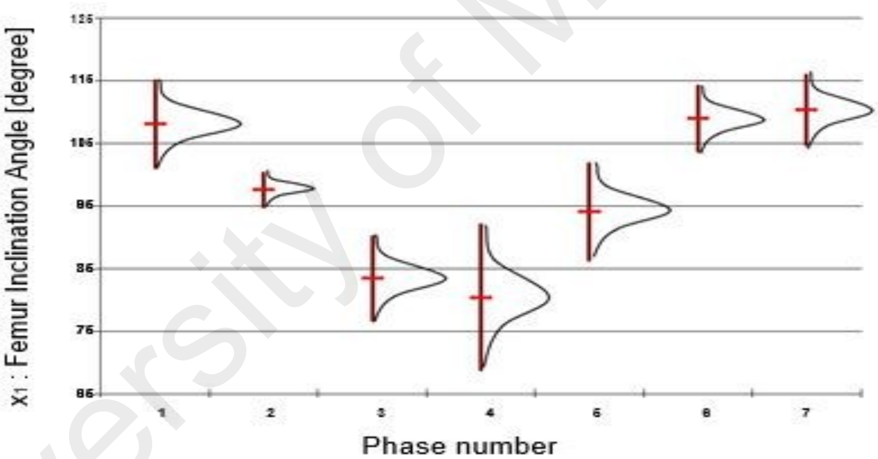


Figure 3.4 : Input membership function for x1

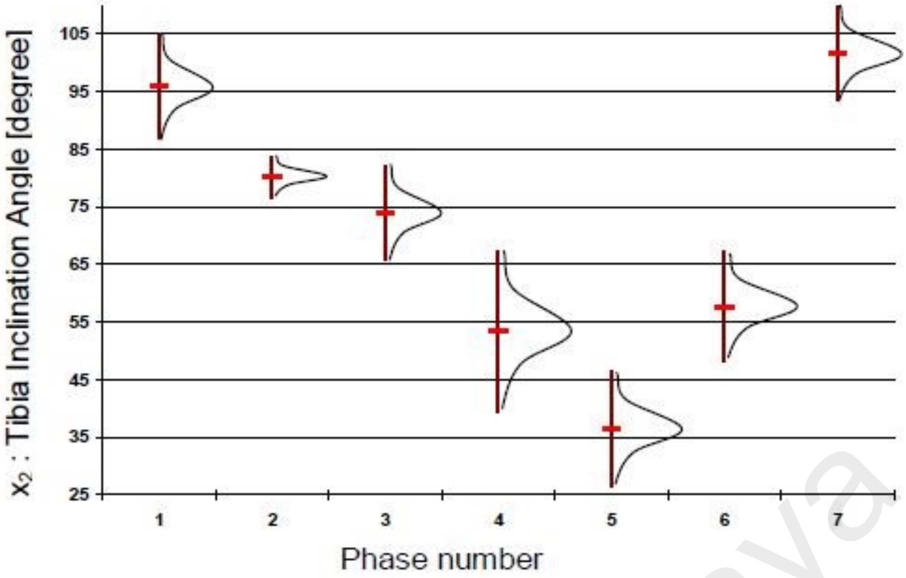


Figure 3.5 : Input membership function for x^2

Table 3-1 : Femur and Tibia angles sample mean of each cluster for each input

Phase	Femur angle (˚)		Tibia angle (˚)	
	Mean	Standard deviation	Mean	Standard deviation
phase 1	108.7	0.801	115.0	5.613
phase 2	97.8	6.182	84.7	4.150
phase 3	78.9	4.908	69.8	6.051
phase 4	76.6	3.409	47.0	6.201
phase 5	97.2	8.084	43.0	6.404
phase 6	113.5	2.069	73.2	10.416
phase7	113.0	1.871	105.9	6.520

Table 3-2 : Knee torque and knee angle sample mean of each cluster for each output

Phase	Knee torque (Nm)		Knee angle (˚)	
	Mean	Standard deviation	Mean	Standard deviation
phase 1	-0.377	21.184	6.26	5.636
phase 2	-19.453	11.403	13.00	2.722
phase 3	4.635	5.815	7.35	2.362
phase 4	-9.487	3.486	30.87	11.123
phase 5	-4.244	2.046	62.33	4.924
phase 6	-2.557	1.652	49.01	9.711
phase7	-10.05	13.677	7.72	11.464

To compare the current state of each input with its last state, the premise will read as;

$$\boldsymbol{if\ x_j(k) is\ Ph_j^i\ AND\ (x_j\ (x-1)\ is\ Ph_j^i\ OR\ x_j(x - 1)\ is\ Ph_j^{i-1}} \quad 3\text{-}11$$

Where *j* is the input index and it can be either 1 or 2, meanwhile *i* represents the gait phase of the input index (*i* = 1,2,…..,7). Ph_j^i is defined as phase fuzzy state *i* for *j* input. The statement above stated that if the current state of input *j* is in phase *i*, then the previous state of that input must be the same phase or either the previous state. Meanwhile, phase number 0 is the same as phase number 7 since the input is iterative.

As discussed above, the input state of x^1 and x^2 must be the same phase at any time. The rules that used a combination of 'AND' and 'OR' connectives can be decomposed into group of rules with just AND connectives for processing using only t-norms. As a result, the two successive rules of the new ANFIS can be expressed as;

$$R_l : if\ x_1(k) is\ Ph_1^i\ AND\ (x_1\ (x-1)\ is\ Ph_1^i\ AND$$

$$\boldsymbol{x_2(k) is\ Ph_2^i\ AND\ (x_2\ (x - 1)\ is\ Ph_2^{i-1}} \quad 3\text{-}12$$

$$THEN\ y_1^l = c_{1,0}^l + c_{1,1}^l x_1 + c_{1,2}^l x_2 + c_{1,3}^l x_3 + c_{1,4}^l x_4$$

$$\boldsymbol{y_2^l = c_{2,0}^l + c_{2,1}^l x_1 + c_{2,2}^l x_2 + c_{2,3}^l x_3 + c_{2,4}^l x_4} \quad 3\text{-}13$$

$$R_{l+1} : if\ x_1(k) is\ Ph_1^i\ AND\ (x_1\ (x-1)\ is\ Ph_1^i\ AND$$

$$\boldsymbol{x_2(k) is\ Ph_2^i\ AND\ (x_2\ (x - 1)\ is\ Ph_2^i} \quad 3\text{-}14$$

$$THEN\ y_1^{l+1} = c_{1,0}^{l+1} + c_{1,1}^{l+1} x_1 + c_{1,2}^{l+1} x_2 + c_{1,3}^{l+1} x_3 + c_{1,4}^{l+1} x_4$$

$$\boldsymbol{y_2^{l+1} = c_{2,0}^{l+1} + c_{2,1}^{l+1} x_1 + c_{2,2}^{l+1} x_2 + c_{2,3}^{l+1} x_3 + c_{2,4}^{l+1} x_4} \quad 3\text{-}15$$

Where y_1 represents the first output of the control input system of knee angle, whereas y_2 represents the second output of the control input system of knee torque. The above statements stated that l represents the rule number, where (l and l+1) = (1 and 2), (3 and 4), …, (13 and 14). Meanwhile, l_{th} $c_{1,0}^l$,…., $c_{1,4}^l$ and $c_{2,0}^l$,…., $c_{2,4}^l$ represents the consequent parameters for the first and second output, respectively $x_j(k)$ and $x_j\ (x-1)$ can be considered as separated inputs that are associated with the same variable. . Hence, the membership functions related to $x_j(k)$ and $x_j\ (x-1)$ are the same. Consequently, the ANFIS control input system consists of four inputs, 14 membership functions and 14 rules. Figure 3.6 and figure 3.7 showed the relationships off all rules and its membership functions.

As discussed above, the proposed fuzzy system was considered as two separated ANFIS system for each output and they were trained individually even though they have an identical input membership function. After training, both systems were considered as one ANFIS with two inputs and two outputs.

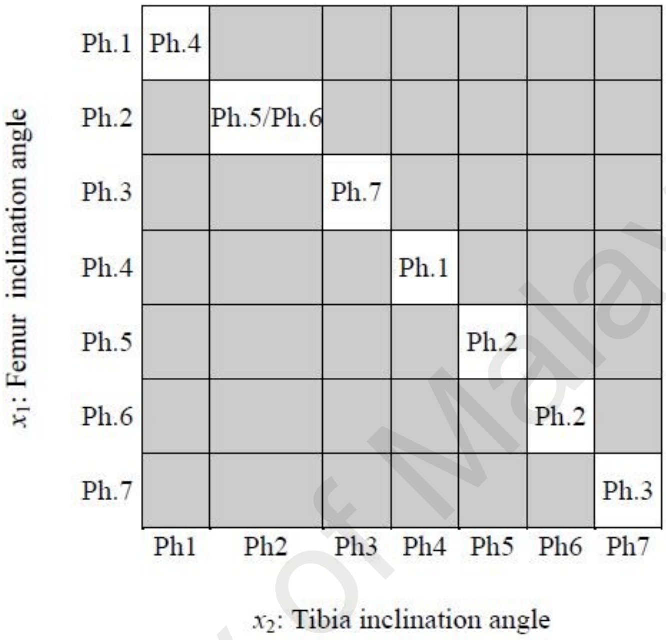


Figure 3.6 : Rotational matrix with the rules of the prosthetic leg controller

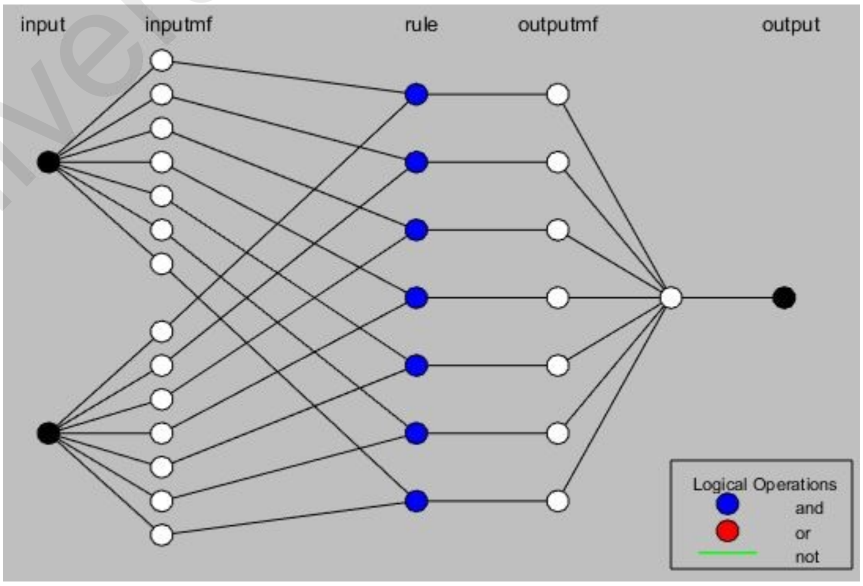


Figure 3.7 : ANFIS model structure

3.3 Validation of an ANFIS-based control input framework of transfemoral prosthetic leg using experimental in- socket sensor system as an input: walking at different speeds

3.3.1 Experimental protocol with transfemoral amputees

In order to evaluate the performance of the control framework of transfemoral prosthetic leg using an in-socket sensory system, a socket with the instrumented piezoelectric sensors (Jasni et al., 2016) was fabricated to be in good fit with the amputee's stump. Eight sensors were embedded onto the anterior wall of the socket while seven sensors embedded on the posterior wall (Jasni et al., 2016) (Figure 3.8). All the sensors were connected to the NI-6009 DAQ CARD for data acquisition process (Figure 3.9). One healthy male transfemoral amputee, 36 years old, participated in the study to acquire walking signals from the in-socket sensory system (Figure 3.10). The amputee was asked to walk on the treadmill set with various speed starting from 0 m.s^{-1} to 2 m.s^{-1} and the speed was gradually decreased to a stop.

A total of 10 complete trials for 3 separate sessions of different days were recorded for further analysis. A switch was placed on the treadmill to synchronize the starting point of all data collection. The output signal from the piezoelectric sensor will be utilized to control the actuated knee of an active prosthetic leg. In order to do that, the signals need to be classify into different motion type via fuzzy classifier method. In this study, the input signal from piezoelectric sensors will be classify into seven gait state which divided into two gait phase which are stance and swing phase. The classified signal will be used as an input for the ANFIS control input system.

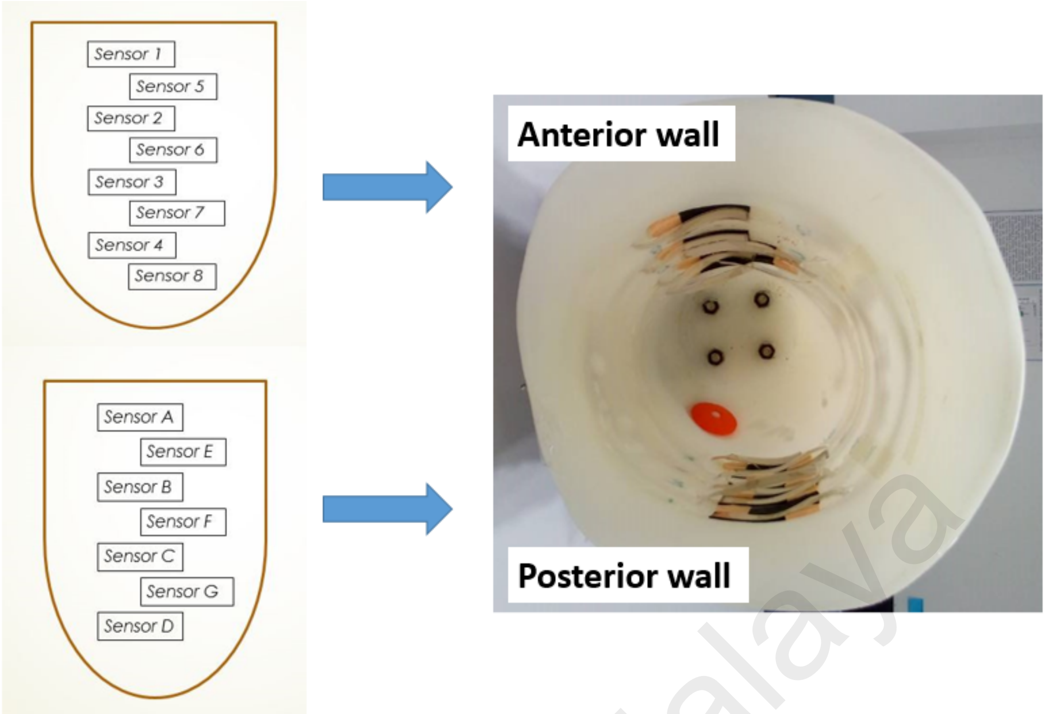


Figure 3.8 : Piezoelectric sensor placement for in-socket sensor system

Figure 3.9 : Socket fitting and alignment for gait trial

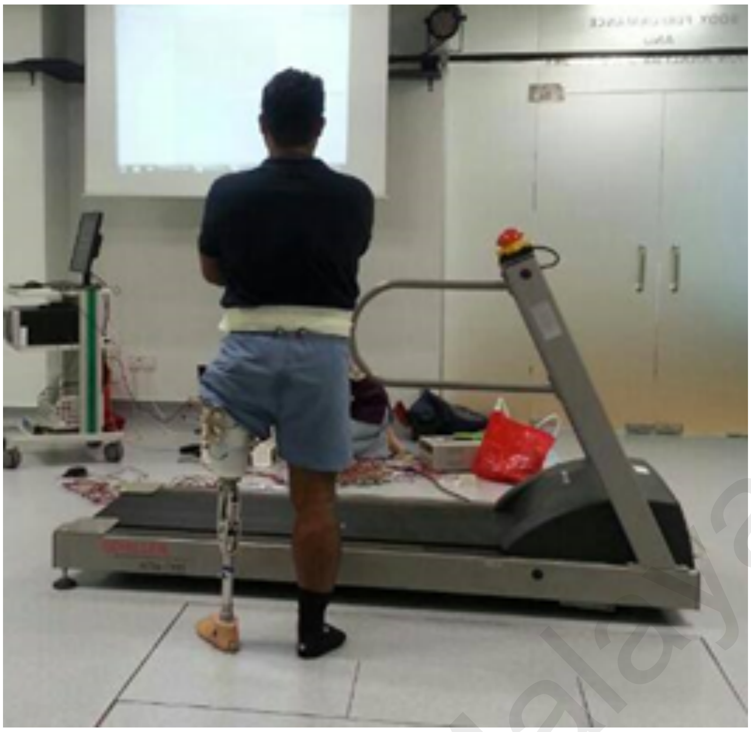


Figure 3.10 : Gait trial experiment

3.3.2 Conversion/mapping process from in-socket sensory system data to inclination angle; i.e. femur and tibia angle for ANFIS control input framework

The output data from the in-socket sensory system was classified and fed into the ANFIS control input in the form of femur and tibia angle through conversion process as shown below (Figure 3.11). The conversion process was done through mapping from in-socket sensory system data to the Vicon nexus system data using MATLAB software for one complete gait cycle. Calculation of inclination angle (femur and tibia angle) were done, since Vicon nexus couldn't provide the data in the form of inclination angle; i.e. tibia and femur angle to fed into the ANFIS control input system and also the in-socket sensory signal output is in the form of voltage. The calculation to find the inclination angle; i.e. femur angle and tibia angle were discussed in the inclination angle section; *section 3.2.1*, as mentioned above.

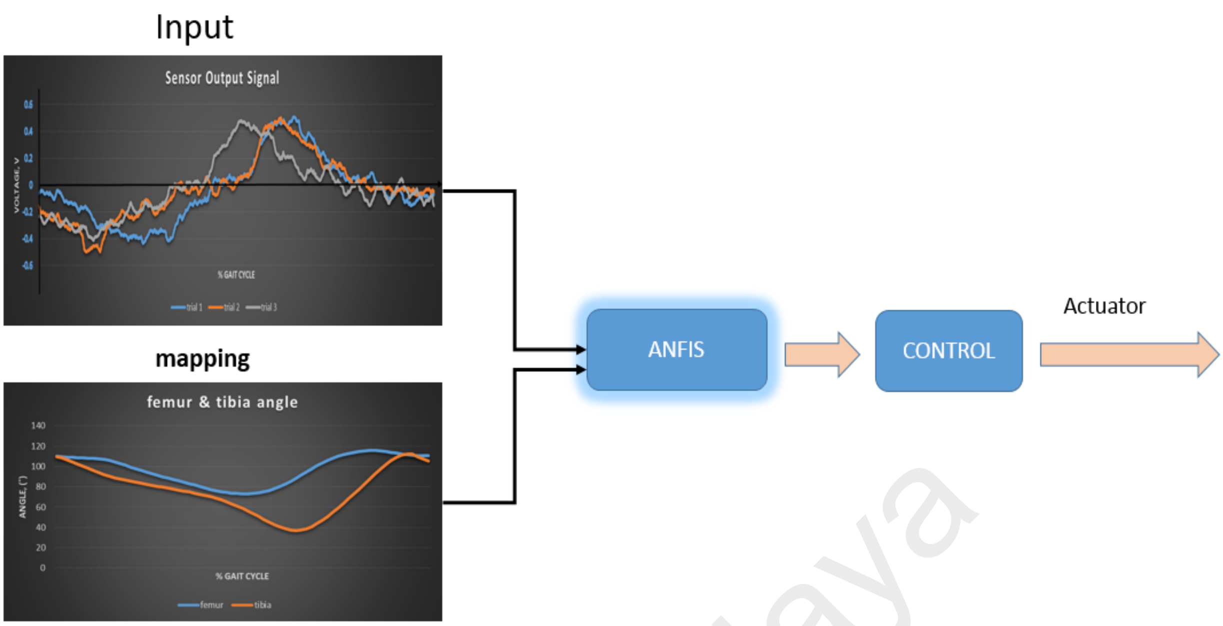


Figure 3.11 : In-socket sensory system mapping for ANFIS control input system

3.3.3 Validation of and ANFIS-based control input framework of transfemoral prosthetic leg using in-socket sensory system

In order to evaluate the performance of the ANFIS with in-socket sensory signal as the input, a reference pattern of the subject's walking gait was applied as an input to the ANFIS system. For further analysis, the knee parameters, i.e. angle and torque were tested under different time bases derived from the different walking speeds. The purpose of using different time bases was to check the fuzzy performance of the system at various speeds. The ANFIS system was further tested with in-socket sensory signals output under different walking speeds starting from 0 m.s^{-1} to 2 m.s^{-1}. The output knee angles and knee torque were obtained and compared with the input signal to verify the ANFIS algorithm.

A set of 180 data points was used as training data to train the algorithm according to its requirements for gait data analysis meanwhile 180 others were used as testing data to verify the ANFIS algorithm for 0 m.s^{-1} to 1 m.s^{-1}. All the 180 set of data represents for one whole gait cycle of normal human walking and the algorithm are the same as discussed above. Meanwhile, for the speed 1 m.s^{-1} to 2 m.s^{-1}, total set of training data and testing data used was the same as speed 0 m.s^{-1} to 1 m.s^{-1}. Lastly, for the speed 2 m.s^{-1} to 1 m.s^{-1} the total set of training and testing data used for simulation was 90 data. The total set of data used are depends gait cycle of a person and also its walking speed since the different speed contribute to different amount of step length during walking. The effectiveness of the ANFIS control input system were calculated by using percentage of accuracy as depicted below for both knee parameters;

Knee angle percentage of accuracy;

$$\boldsymbol{percentage\ of\ accuracy};\% = \left\{\frac{\mathbf{1}}{\boldsymbol{n}}\sum_{\boldsymbol{n=1}}^{\boldsymbol{n=j}} \mathbf{1} - \left\|\left[\frac{\boldsymbol{\theta_{ANFIS}} - \boldsymbol{\theta_{ref}}}{\boldsymbol{\theta_{ref}}}\right]\right\|\right\} \times \mathbf{100}\% \tag{3-16}$$

Knee torque percentage of accuracy;

$$\boldsymbol{percentage\ of\ accuracy};\% = \left\{\frac{\mathbf{1}}{\boldsymbol{n}}\sum_{\boldsymbol{n=1}}^{\boldsymbol{n=j}} \mathbf{1} - \left\|\left[\frac{\boldsymbol{\tau_{ANFIS}} - \boldsymbol{\tau_{ref}}}{\boldsymbol{\tau_{ref}}}\right]\right\|\right\} \times \mathbf{100}\% \tag{3-17}$$

Where n is the total number of data used in the ANFIS control input system, and j is the last data fed into the ANFIS control input system.

CHAPTER 4: RESULTS AND DISCUSSION

4.1 Implementation of an ANFIS for control input framework of an active prosthetic leg

The FIS outputs for the training data are depicted in figure 4.1 and figure 4.2. Meanwhile, figure 4.3 and figure 4.4 showed surface viewer for whole ANFIS system using sub-cluster method.

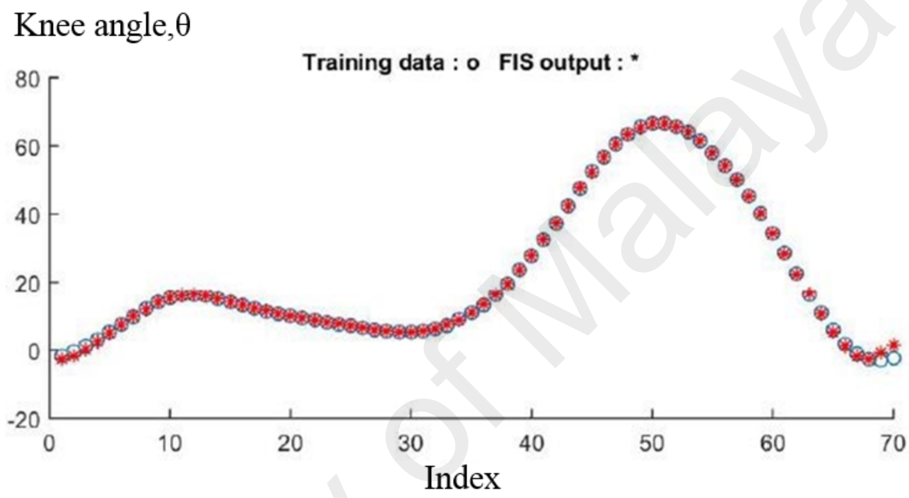


Figure 4.1 : ANFIS output angle versus training data

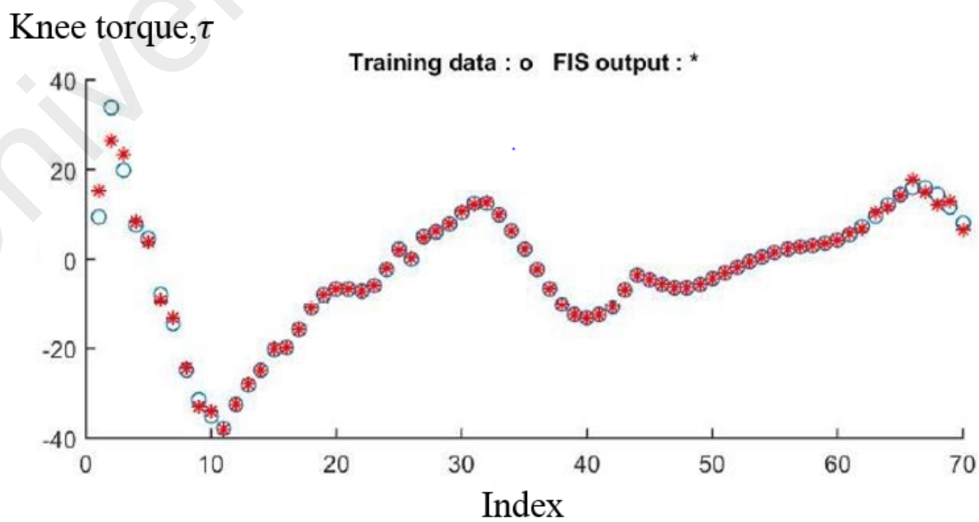


Figure 4.2 : ANFIS torque output versus training data

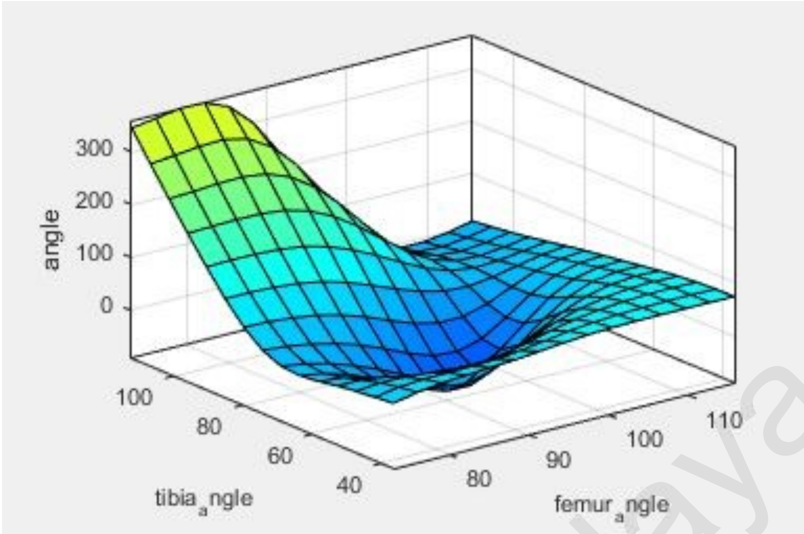


Figure 4.3 : Surface viewer of ANFIS for knee angle output

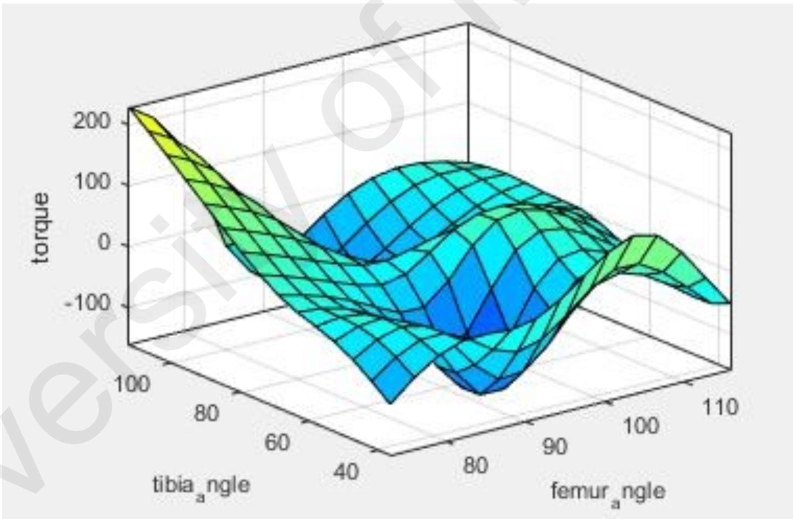


Figure 4.4 : Surface viewer of ANFIS for knee torque output

Results depicted in the Figure 4.1 and 4.2 showed that the knee angle and knee torque pattern of the output can followed closely with the reference pattern set in the ANFIS control input system. Thus, it proved that the developed algorithm can be used for an active prosthetic leg. In addition, the percentage of accuracy of the ANFIS control input system when tested with the Winter data was 99% of accuracy.

4.2 Validation of an ANFIS-based control input framework of transfemoral prosthetic leg using experimental in- socket sensor system as an input: walking at different speeds

4.2.1 Experimental trial with a transfemoral amputee wearing an in-socket sensory system

The purpose of using piezoelectric as sensory system for the transfemoral socket was because of piezoelectric were able to detect the changes in muscle contraction level or force. Thus, the location to place the sensor was very crucial and has reported in any other by the other study (Jasni et al., 2016). In order to do that, the quadriceps and hamstring muscle group were targeted. For analysis purpose, only two sensors were chosen for both muscle group, since the sensors were located on the biggest area of the stump and was identified as the most crucial muscle used for flexion and extension of the knee during walking. Moreover, both have strong contact between the stump and sensor and hence can be measure easily. Figure 4.5 and Figure 4.6 showed the output voltage signal for the sensor that was placed on the anterior side of the stump (quadriceps).

Meanwhile, Figure 4.8 and Figure 4.9 showed the output voltage signal for the sensor place on the posterior side of the stump (hamstring). The results (Figure 4.8 and 4.9) proved that all the sensors were working perfectly in the designated position to detect pressure or force changes of the muscles, while the patient was walking regardless of the varying walking speeds. The consistency of output signals by the sensor from each trial implied the efficacy of the sensory system to be used as control input for the active prosthetic knee. In addition, Figure 4.7 summarized the total 10 trials for both sensor B and sensor C in the form of mean voltage sensor output signal. While, Figure 4.10 summarized the total 10 trials for both sensor 2 and sensor 3 in the form of mean voltage sensor output signal.

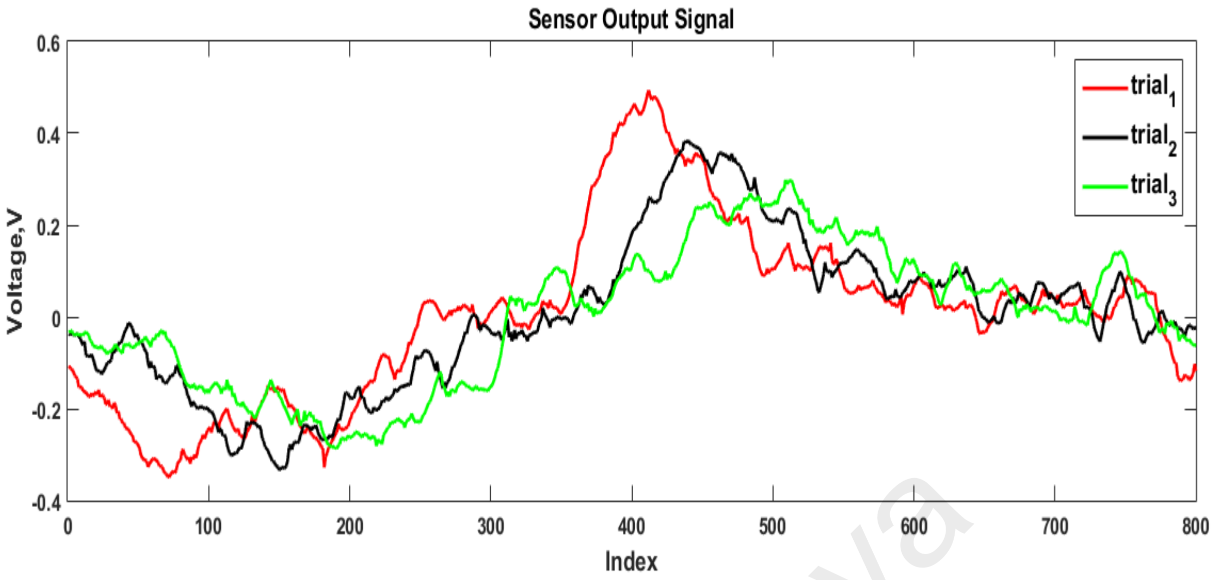


Figure 4.5 : Output voltage signal of sensor B (anterior) for three trial during one complete gait cycle, 100%

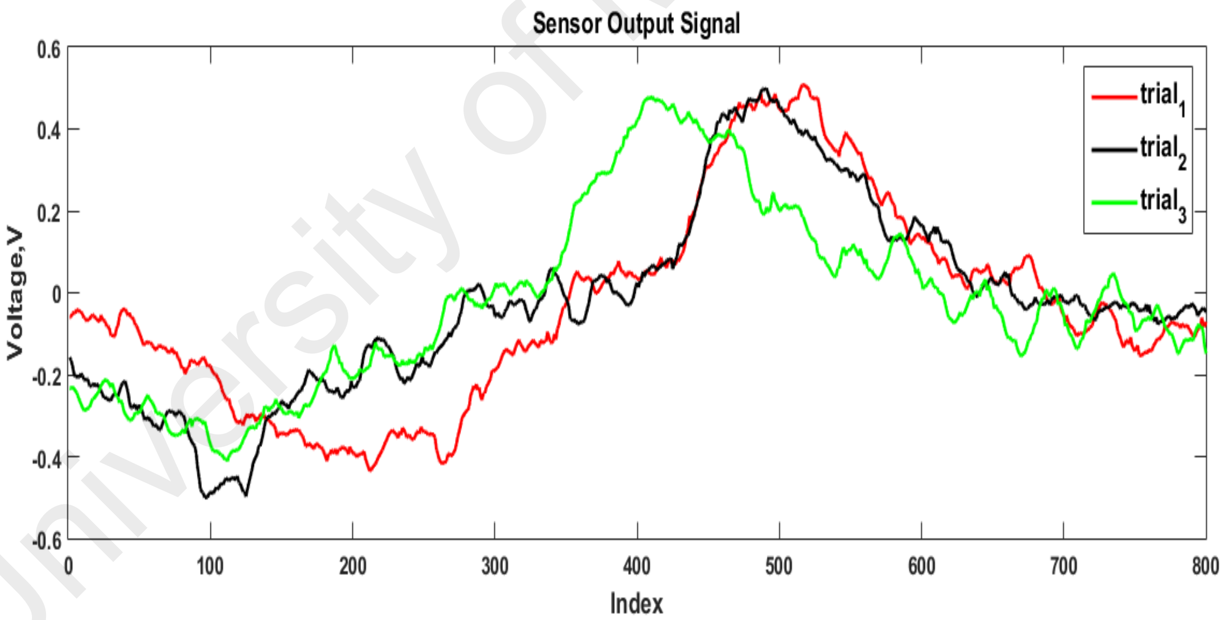


Figure 4.6 : Output voltage signal of sensor C (anterior) for three trial during one complete gait cycle, 100%

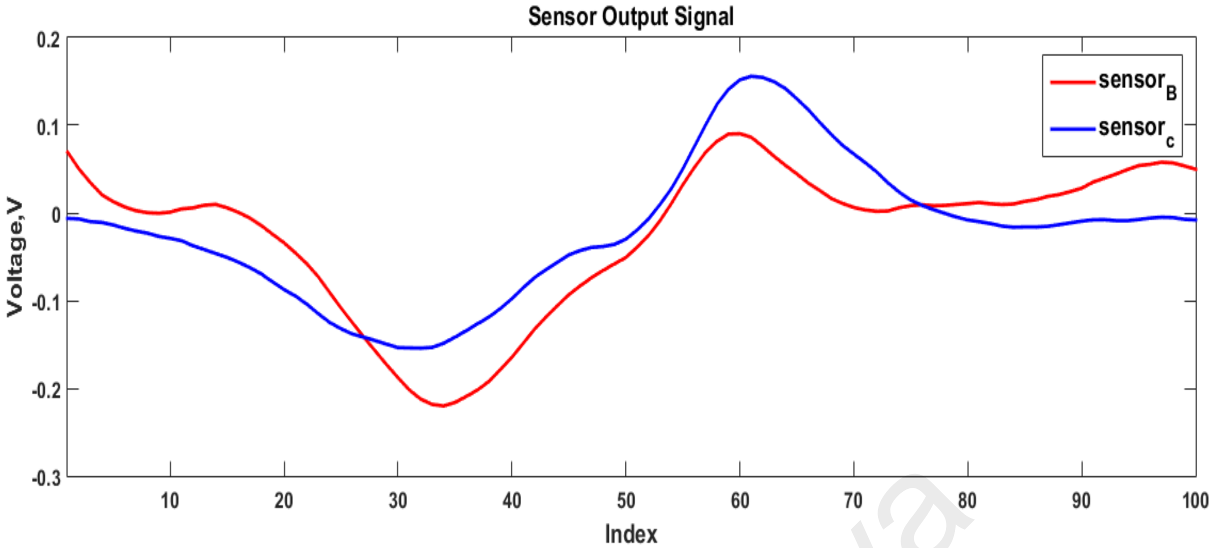


Figure 4.7 : Mean of voltage output signal sensor B and C for 10 trials during one complete gait cycle, 100%

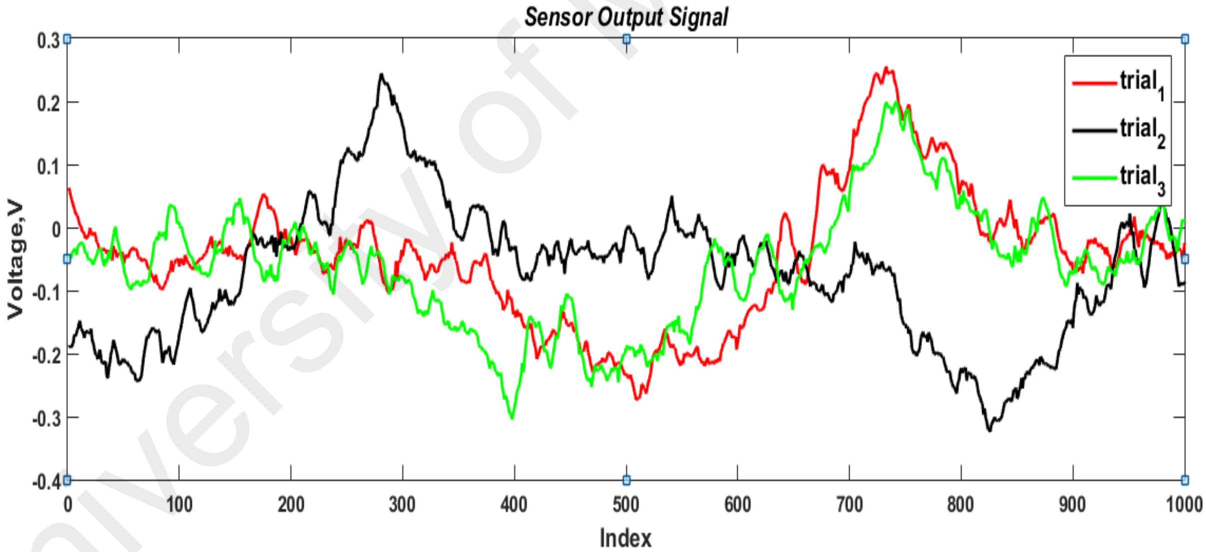


Figure 4.8 : Output voltage signal of sensor 2 (posterior) for three trials during one complete gait cycle, 100%

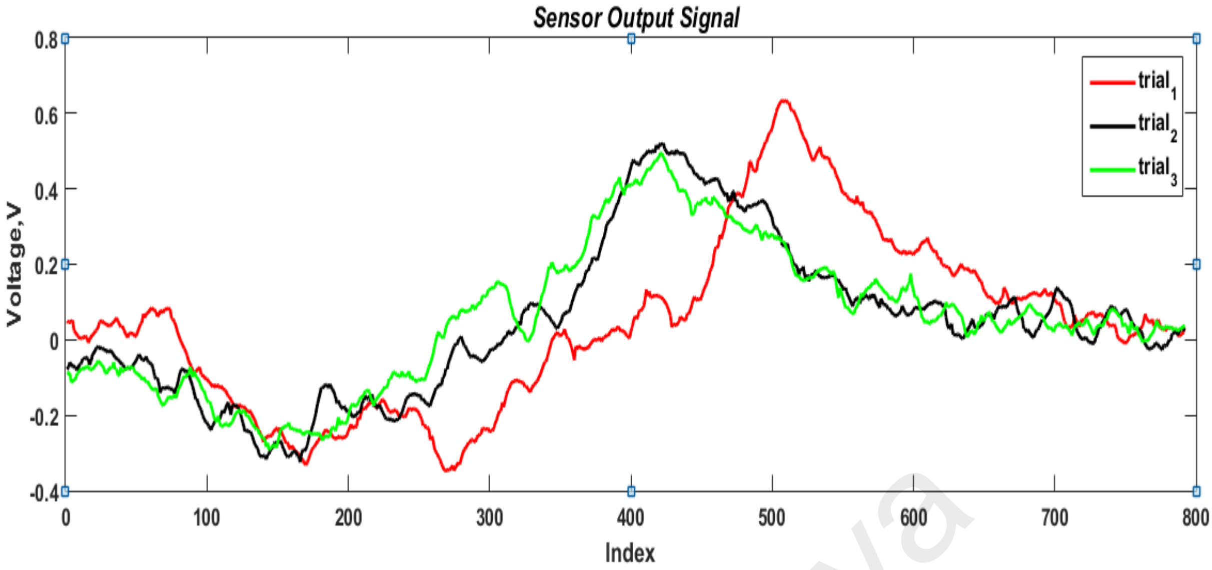


Figure 4.9 : Output voltage signal of sensor 3 (posterior) for three trials during one complete gait cycle

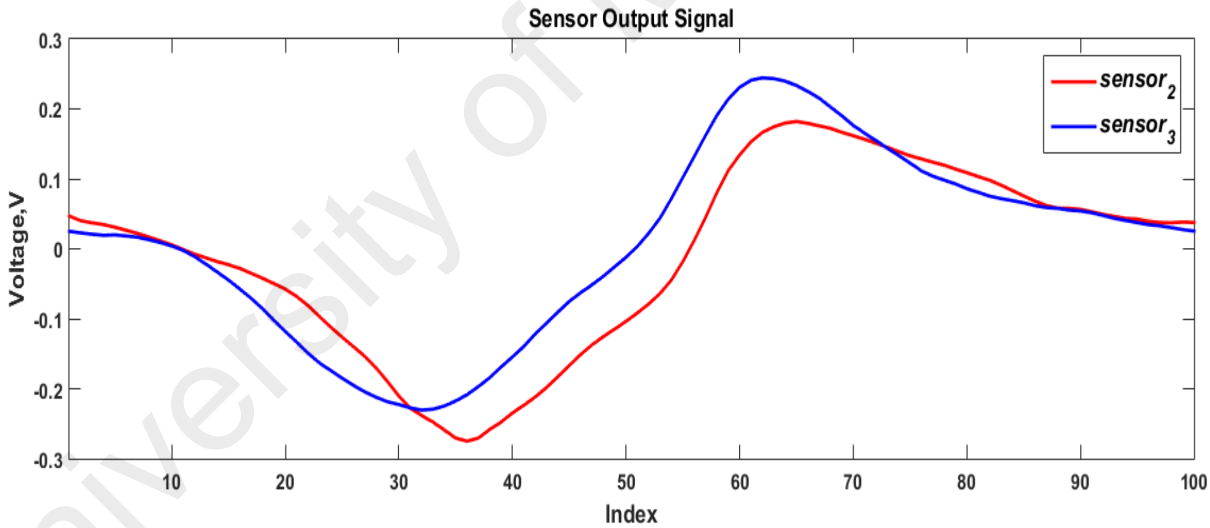
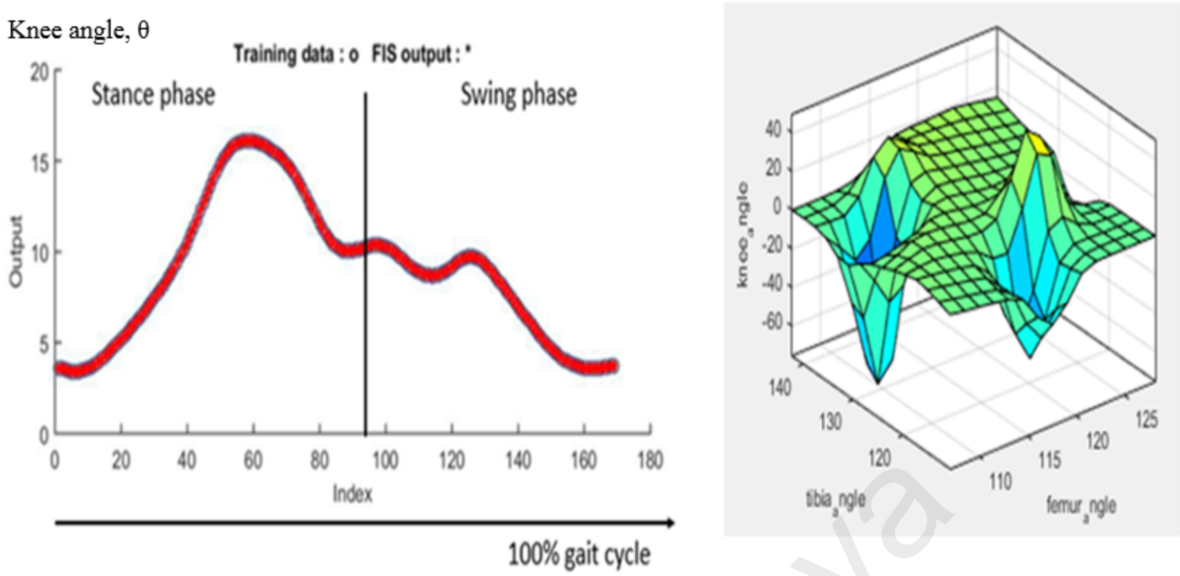


Figure 4.10 : Mean of voltage output signal sensor 2 and 3 for 10 trials during one complete gait cycle, 100%

4.2.2 Validation of the control framework of transfemoral prosthetic leg using in-socket sensory system

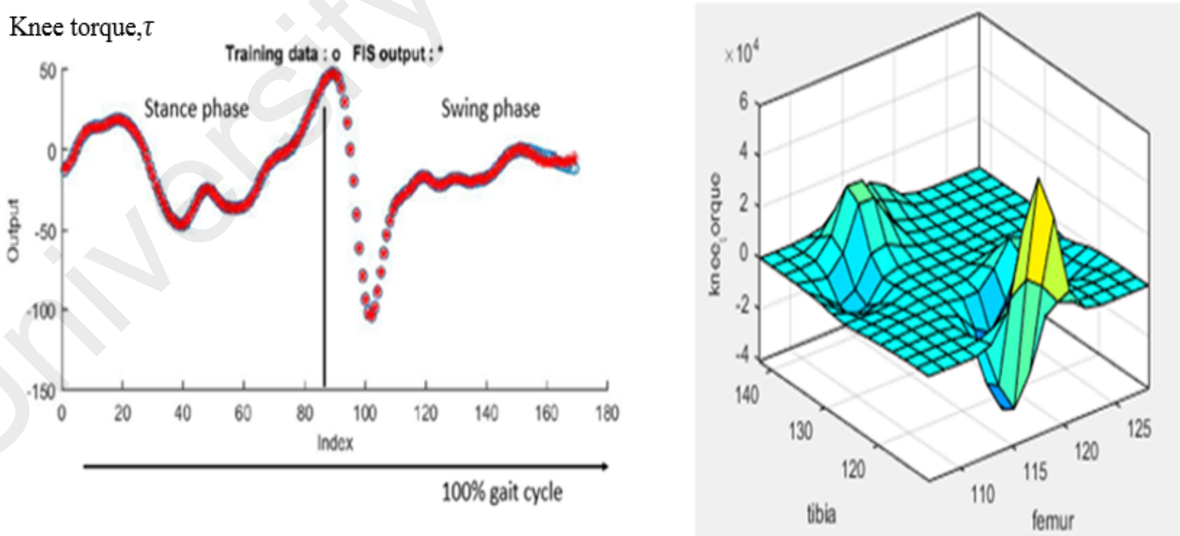
Figure 4.7 and Figure 4.8 showed the ANFIS output knee angle and knee torque during walking speed 0 ms^{-1} to 1 ms^{-1} for one complete gait cycle. As shown in the results, the output of knee angle and knee torque obtained proved that the ANFIS system can be used as algorithm for the in-socket sensory system. As illustrated in the Figure 4.7 and 4.8, testing data used as an input for the controller followed the pattern of the training data fed into the ANFIS system to trace the gait cycle. In addition, percentage of accuracy, % during this walking speed shows 95% of accuracy for both knee angle and knee torque output when compare the reference pattern with the output of the ANFIS system. Thus, this verified that the ANFIS algorithm could be used for the in-socket sensory system of an active prosthetic leg system in different locomotion styles at various walking speeds.

Meanwhile, Figure 4.9 and 4.10 showed the results of ANFIS output knee angle and knee torque during walking speed 1 ms^{-1} to 2 ms^{-1} for one complete gait cycle. At this change of walking speed, the ANFIS output also showed that knee angle and knee torque pattern of normal walking can be achieved through the ANFIS algorithm. In addition, percentage of accuracy, % during this walking speed shows 95% of accuracy for knee angle output while 85% for knee torque output when compare with the reference pattern. Other than that, Figure 4.11 and Figure 4.12 showed the results of ANFIS output knee angle and knee torque during walking speed 2 ms^{-1} to 1 ms^{-1} for one complete gait cycle. At this state the ANFIS output for both knee angle and knee torque pattern could follow normal human walking gait. In addition, percentage of accuracy, % during this walking speed shows 95% of accuracy for knee angle output while 93% for knee torque output when compare the reference pattern with the output of the ANFIS system This further verified the versatility of the ANFIS algorithm.



(a) (b)

Figure 4.11 : (a) ANFIS output knee angle versus training data for gait cycle, 0 m.s^{-1} to 1 m.s^{-1}; (b) ANFIS for knee angle output versus training data for gait cycle, 0 m.s^{-1} to 1 m.s^{-1}



(a) (b)

Figure 4.12 : (a) ANFIS output knee torque versus training data for gait cycle, 0 m.s^{-1} to 1 m.s^{-1}; (b) ANFIS for knee angle output versus training data for gait cycle, 0 m.s^{-1} to 1 m.s^{-1}

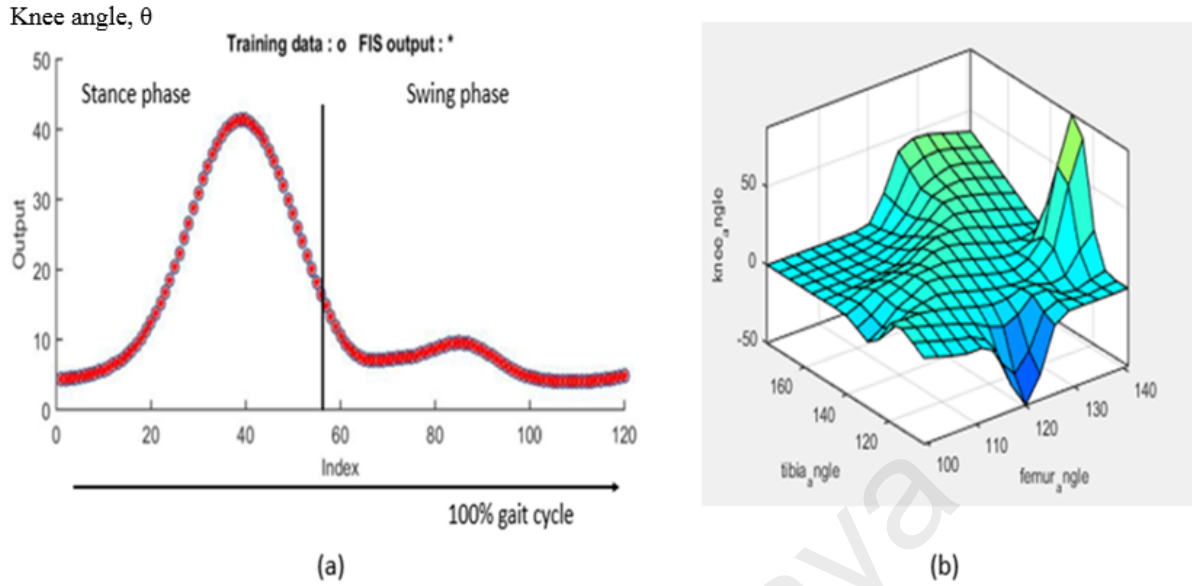


Figure 4.13 : (a) ANFIS output knee angle versus training data for gait cycle, 1 m.s^{-1} to 2 m.s^{-1}; (b) ANFIS for knee angle output versus training data for gait cycle, 1 m.s^{-1} to 2 m.s^{-1}

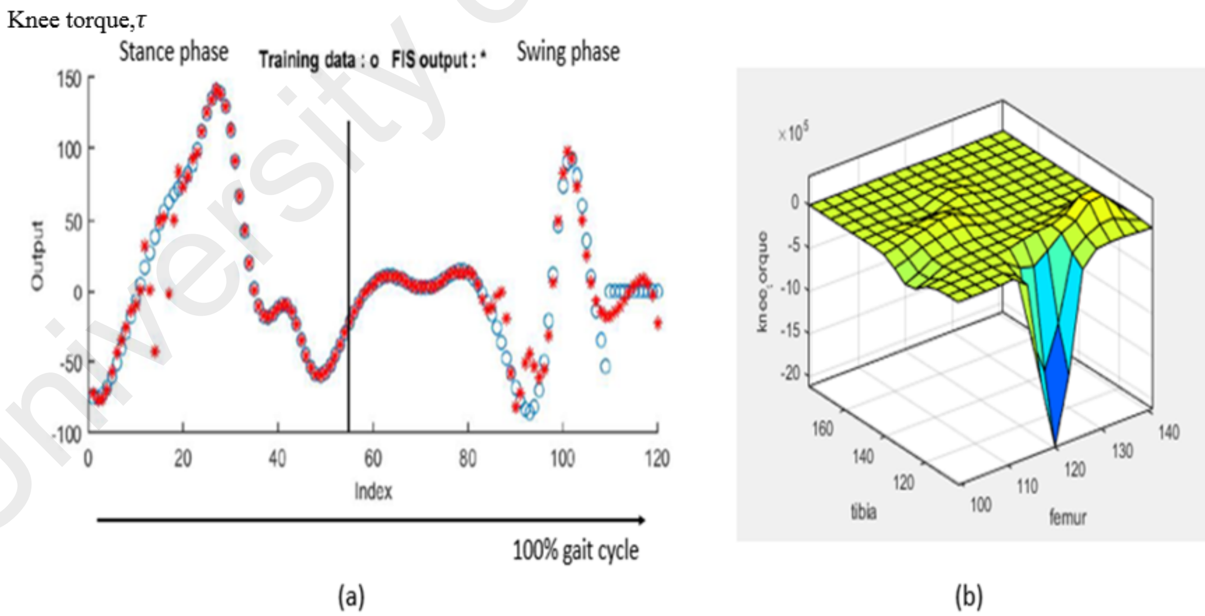


Figure 4.14 : (a) ANFIS output knee torque versus training data for gait cycle, 1 m.s^{-1} to 2 m.s^{-1}; (b) ANFIS for knee torque output versus training data for gait cycle, 1 m.s^{-1} to 2 m.s^{-1}

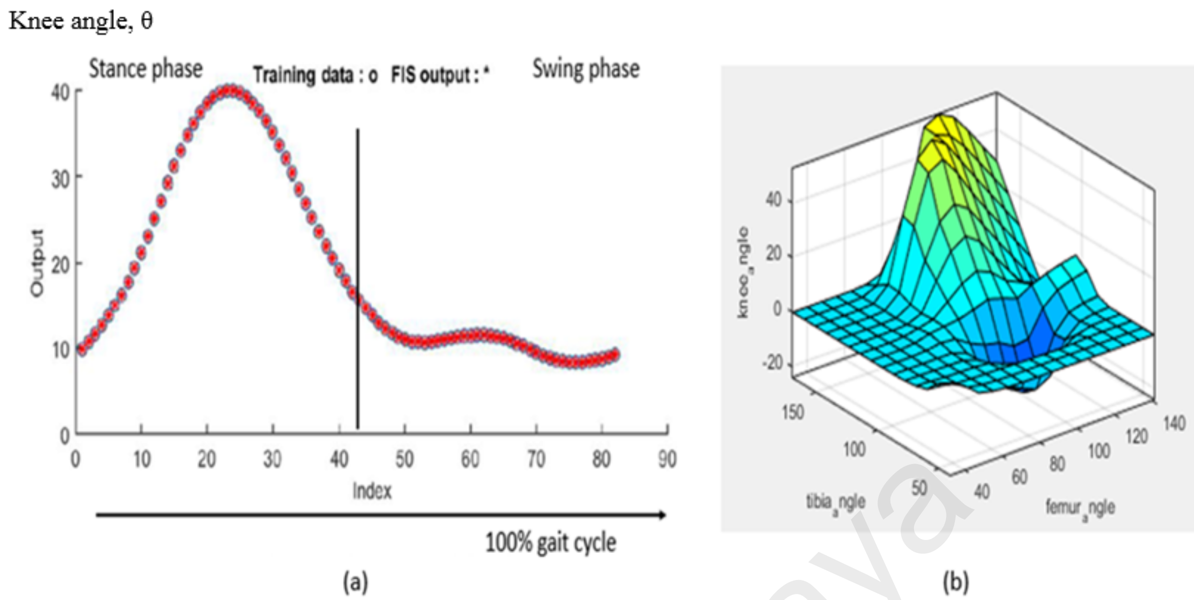


Figure 4.15 : (a) ANFIS output knee angle versus training data for gait cycle, 2 m.s^{-1} to 1 m.s^{-1}; (b) ANFIS for knee angle output versus training data for gait cycle, 2 m.s^{-1} to 1 m.s^{-1}

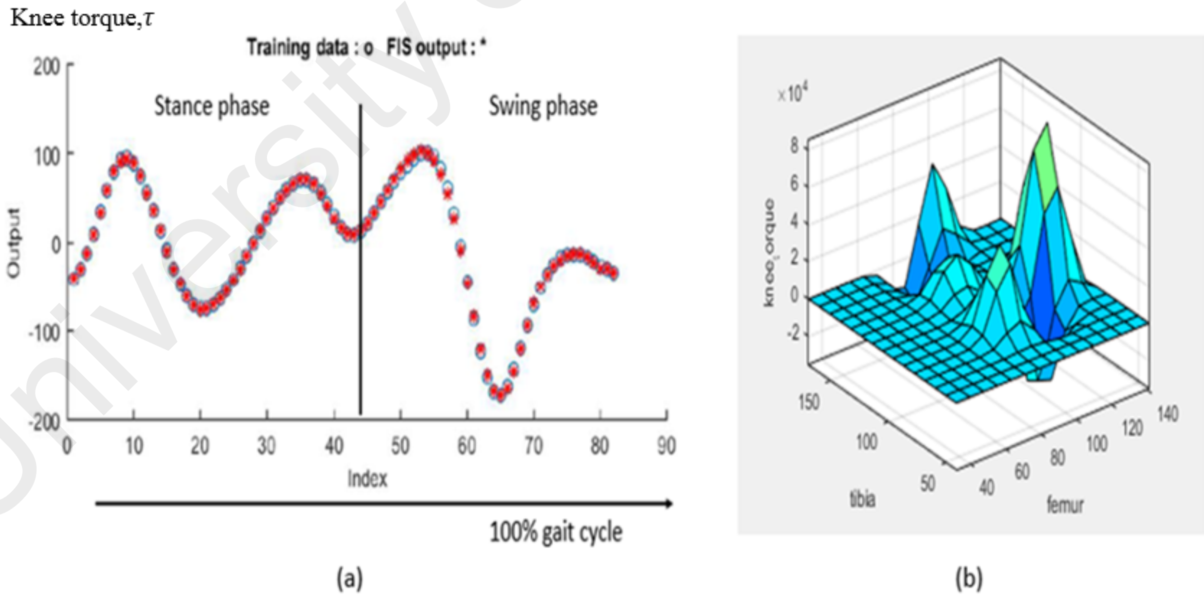


Figure 4.16 : (a) ANFIS output knee torque versus training data for gait cycle, 2 m.s^{-1} to 1 m.s^{-1}; (b) ANFIS for knee torque output versus training data for gait cycle, 2 m.s^{-1} to 1 m.s^{-1}

Table 4-1 : Accuracy of knee parameter from ANFIS control input system between Winter data and experimental data with in-socket sensory system

Knee parameters	*Winter data (%)*	*Experimental data (%)*
Knee angle	99	99
Knee torque	99	93

Table 4-2 : Accuracy of knee parameter from ANFIS input control system using experimental data with in-socket sensory system at different walking speeds

Knee parameters	*0 to 1 ($m.s^{-1}$) (%)*	*1 to 2 ($m.s^{-1}$) (%)*	*2 to 1 ($m.s^{-1}$) (%)*
Knee angle	99	99	99
Knee torque	96	80	95

Despite that, the uncertainty in joint angles measurements due to the different patterns of normal walking cycle between individual or even day-to-day activity of the same subject (inter-and intra-subject) need to be considered in designing the knowledge-based system. The uncertainty of input data corrupted by measurement noise was discussed very briefly. Some of the noise was due to missing marker during experimental session. Thus, the data cannot be captured clearly during for the walking data session. However, due to the uncertainty of definition of slow, normal, and fast walking, uncertainty of Gaussian input membership functions must be considered to increase the robustness of the model. This leads to uncertainty about determining the membership value of the entered data (type-2 FIS). Therefore, further study is needed to develop a type-2 FIS for the actuated prosthetic knee controller.

CHAPTER 5: CONCLUSION

This thesis presented feasibility study towards an active prosthetic leg that could potentially enable people to perform different type of motions that require power in lower limb joints such as slope walking or stairs climbing. The signal classification criteria developed using ANFIS based on Winter's databased proved to be efficient to be used on the active prosthetic tested with the Winter data. It showed almost 99% of accuracy for both knee parameters, knee angle and also knee torque. Thus, it proved that the seven phase of gait state perfectly synchronized with the normal human walking pattern and can be used in any other circumstances.

Moreover, ANFIS as a control input framework was confirmed to be viable with an in-socket sensory system for an active prosthetic leg. Signal classification criteria developed by using ANFIS could be used to accurately classify different gait cycles and assign accurate knee angle and torque as output in each gait phase. This result strongly suggested that ANFIS was suitable and useful for an active prosthetic knee. The seven phases of the gait state synchronized with the normal human walking pattern. Thus, the ANFIS system could be used at different increasing and decreasing walking speeds. From the ANFIS control input system, almost 95% accuracy for the knee angle output pattern and 95% accuracy for the knee torque output pattern were detected in various tested speeds. Thus, the control input system for an ANFIS could enable prosthetic legs to replicate human motion ambulation at various speeds and conditions.

Further trials with the amputees should be conducted to determine the ANFIS control input performance in other locomotion settings, such as stair climbing or sit-to-stand movement. In addition, other data analyses should be performed with different amputees and amputated stump types because every human has different gait pattern and anatomical stump shapes. Thus, the effectiveness of the ANFIS input control system on different patterns of human walking gait and stumps can be verified. In conclusion, the ANFIS algorithm presented in this study could derive user intention from the in-socket sensory system by providing the pattern recognition of amputees' walking gait. This method potentially helped to reduce the metabolic energy rate consumption and the gait deviation problem of amputees' user. More field tests with prosthetic users in daily life are planned in the near future to further optimized the control algorithm.

The social and economic benefits from this new development are obvious. This ANFIS-based control algorithm is optimally adapted to the user's needs and flexible to cope with all circumstances the user encounters. It surely enhances the amputee's quality of life and their integration in normal life.

5.1 Future work

Further trials with the amputees need to be done to determine the ANFIS control input performance in other locomotion settings such as climb upstairs, slope, and sit to stand. In addition, more data analysis can be done with different types of amputees since every human has different walking gait pattern. Thus, it can also verify the effectiveness of the ANFIS input control system on different patterns of human walking gait. Moreover, various type of transfemoral socket could be tested too since different type of socket exerted different amount of pressure to the stump.

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LIST OF PUBLICATIONS AND PAPERS PRESENTED

Conference

1. **Nur Hidayah Mohd Yusof**, Farahiyah Jasni, Nur Azah Hamzaid, Khin Wee Lai. Feasibility study of microcontroller knee joint for an active transfemoral prosthetic leg based-in socket sensor (piezoelectric). ISPO 16th World Congress in Cape town, Accepted for publication on May, 2017
2. **Nur Hidayah Mohd Yusof**, Farahiyah Jasni, Nur Azah Hamzaid, Khin Wee Lai. Adaptive network based fuzzy inference system (ANFIS) for an active transfemoral prosthetic leg using in-socket sensory system. Conference for Innovation in Biomedical Engineering and Life Sciences (ICIBEL) 2017. Accepted for oral presentation.

Journal paper

1. **Nur Hidayah Mohd Yusof**, Farahiyah Jasni, Nur Azah Hamzaid, Khin Wee Lai. In-Socket Sensory System with Adaptive Neuro Based Fuzzy Inference System for Active Transfemoral Prosthetic Leg, Journal of Electronic Imaging, accepted for publication October, 2018.