DESIGN OF KNEE JOINT MECHANISMS AND IN-SOCKET SENSORS FOR TRANSFEMORAL AMPUTEES

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FACULTY OF ENGINEERING
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KUALA LUMPUR

2015

DEDICATION

To my beloved Mother, my Wife for her endless support, my Sons (Youssef and Adam) and my Daughter (Salma).

To the spirit of the late Father.

UNIVERSITI MALAYA

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Title of Project Paper/Research Report/Dissertation/Thesis ("this Work"):

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Field of Study: Biomechatronics

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ABSTRACT

Lower limb prostheses are developed to assist amputees in restoring mobility functions such as walking, sit-to-stand, stair ascent/descent, and ramp climbing. Although the current prostheses are equipped with sensors, actuators, controllers, and mechanical structures, they require improvements to mimic the function of the natural limbs. The first challenge in prosthetic development is to monitor the amputee/prosthesis interaction by using sensors built into the socket. This interaction helps in detecting the gait phases and events, in addition to develop new control strategies for prostheses, which may enhance the amputees' comfort. The second challenge is to develop a knee prosthetic mechanism that could imitate the functions of the natural knee.

To accomplish the aims of this thesis, studies were undertaken consecutively. First, the technology of the knee prosthesis was studied to understand the functionality of its components. The technology review showed that the sensory system requires enhancement, in particular, a new sensory system can be added-on to the mechanical sensors to sense the user's intent, identify the transition between phases, and improve the control performance of the prosthesis. Based on this study, the piezoelectric bimorph (PB) was selected as the sensing element while a linear motor was selected as the most appropriate actuator.

Next, the PB was validated as a sensing element by finding out its characteristics for the intended application. The static and dynamic characteristics of the PB were investigated and tested as an in-socket sensor with a transfemoral amputee to check its ability to sense the movement of the knee prosthesis. Moreover, the PB was tested as an actuation element in an application named microgripper that was capable of grasping a small object. Also, the PB was compared with a force sensitive resistor (FSR) as an in-socket sensor for a transfemoral

amputee performing activities such as walking, sit-to-stand, and stair climbing. The PB could track the knee angle at most of the activities, while the FSR could be used as a trigger sensor at different movements.

In the second stage, the focus was on the actuation system and mechanical structure of the knee prosthesis. It was found that, the mechanical actuation system needs improvement in terms of the normal range of motion and the power generation in activities that require extra torque and power. Therefore, a new design of knee prosthesis mechanism that contains a linear actuation system was presented and modeled using a physical modelling tool. The mechanism was physically simulated and controlled using PID controller at activities of daily living (ADL). Finally, an overall control framework of the knee mechanism using in-socket sensor was presented to guide the researchers to develop a knee prosthesis that could be controlled using in-socket sensors.

In conclusion, the study demonstrates the possibility of using the piezoelectric bimorph as an in-socket transducer. Furthermore, a knee prosthesis mechanism was successfully designed, modelled, and tested at ADL. Further, clinical trials are recommended for the knee mechanism upon future development. Moreover, more subjects with different types of sockets may be tested towards improving the functionality of the knee prosthesis.

ABSTRAK

Prostesis betis dibangunkan untuk membantu amputi dalam mengembalikan fungsi pergerakan seperti berjalan, duduk-ke-berdiri, naik turun tangga, dan ber jalan mendaki cerun. Walaupun alatan prostesis yang ada sekarang dilengkapi dengan alat pengesan, penggerak, pengawal, dan struktur mekanikal, alatan prosthesis ini memerlukan penambahbaikan bagi meniru fungsi anggota badan semula jadi. Cabaran pertama dalam pembangunan prostesis adalah dalam memantau interaksi antara amputi dengan alatan prostesis yang dipakainya dengan menggunakan alat pengesan yang dibina di dalam soket. Interaksi ini membantu dalam mengesan fasa dan gaya berjalan, di samping membangunkan strategi kawalan baru untuk prostesis, yang boleh meningkatkan keselesaan amputi tersebut. Cabaran kedua adalah dalam membangunkan satu mekanisme prostesis lutut yang boleh meniru fungsi lutut semula jadi.

Untuk mencapai matlamat tesis ini, beberapa kajian telah dijalankan. Pertama, teknologi prostesis lutut dikaji untuk memahami fungsi komponen-komponennya. Kajian menunjukkan bahawa teknologi sistem deria memerlukan peningkatan, khususnya, sistem deria baru boleh ditambah bersama alat pengesan mekanikal untuk mengesan niat pengguna secara tidak langsung, mengenal pasti peralihan antara fasa pergerakan kaki, dan meningkatkan prestasi kawalan prostesis. Berdasarkan kajian piezoelektrik bimorph (PB) telah dipilih sebagai elemen penderiaan manakala motor linear telah dipilih sebagai penggerak yang paling sesuai.

Seterusnya, PB telah disahkan sebagai elemen penderiaan yang sesuai dengan mengenalpasti ciri-ciri maklumat yang dikehendaki. Ciri-ciri statik dan dinamik PB telah disiasat dan diuji sebagai penderia dalam soket dengan amputi transfemoral untuk memeriksa

keupayaan mengesan pergerakan prostesis lutut. Selain itu, PB telah diuji sebagai satu elemen dalam aplikasi menggerakkan microgripper yang mampu menggenggam objek kecil. Di samping itu, PB telah dibandingkan dengan perintang sensitive daya (FSR) sebagai pengesan dalam soket untuk amputi transfemoral yang melakukan aktiviti seperti berjalan, duduk-ke-berdiri, dan memanjat tangga. PB dapat mengesan sudut lutut untuk kebanyakan aktiviti, manakala FSR boleh digunakan sebagai sensor pencetus pada pergerakan yang berbeza.

Pada peringkat kedua, tumpuan adalah pada sistem penggerak dan struktur mekanikal prostesis lutut. Kajian telah mendapati bahawa, sistem penggerak mekanikal memerlukan peningkatan keperluan mekanikal dari segi julat normal gerakan dan penjanaan tenaga dalam aktiviti yang memerlukan daya kilas dan tenaga yang tinggi. Oleh itu, reka bentuk baru mekanisme prostesis lutut yang mengandungi sistem penggerak linear telah dibentangkan dan dimodelkan menggunakan alat pemodelan fizikal. Mekanisme ini telah disimulasi secara fizikal dan dikawal menggunakan pengawal PID bagi aktiviti harian asas. Akhirnya, satu rangka kerja kawalan keseluruhan mekanisme lutut menggunakan sensor dalam soket disampaikan untuk memandu penyelidik membangunkan prostesis lutut yang boleh dikawal menggunakan sensor dalam soket.

Kesimpulannya, kajian ini jelas menunjukkan kebolehupayaan menggunakan piezoelektrik bimorph sebagai alat pengesan pergerakan kaki yang dimasukkan di dalam soket. Tambahan pula, mekanisme prostesis lutut telah berjaya direka bentuk, dimodel, dan diuji pada aktiviti harian asas. Ujian klinikal juga telah disarankan untuk mekanisme lutut selepas proses pembangunan pada masa depan. Selain itu, kajian dengan lebih ramai pengguna prostetik kaki atas lutut dengan pelbagai jenis soket adalah disarankan ke arah meningkatkan fungsi prostesis lutut tersebut.

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Amr Mohammed, 2015

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CHAPTER 1 : INTRODUCTION

1.1 Introduction

Disability in some people causes limitations in movement, vision, respiration, hearing, and balance. Disabled people have the same health needs as non-disabled people that let them do the activities of daily living (ADL) such as socializing, work, and sports. In particular, a person with limitations in movements, especially people with lower limb amputations, are unable to perform basic daily movements such as walking, running, standing, siting, and stair ascent/ descent without any assistive devices. Movement related disability may be either congenital, i.e. present from birth, or occurs during a person's lifetime. For instance, some people may face some diseases or accidents during their lifetime leading to upper or lower extremities amputations.

The level of lower limb amputations is defined with respect to the knee joint. Thus, below knee amputations (Transtibial) and above knee amputations (Transfemoral) are categorized as the two types of amputations of the lower extremities. People who lost their lower extremities need assistive devices to help them to perform movements. Thus, different classes of assistive devices are used by amputees who suffer from transtibial and transfemoral amputations (Brooker, 2012). In fact, transfemoral amputations are increasing each year, an average of 185,000 amputations are found each year due to accidents and diseases like diabetes and peripheral vascular disease (McGimpsey & Bradford, 2011). Therefore, the need to restore mobility to amputees is necessary especially to those doing repetitive daily activities. The assistive devices and prostheses are used by the amputees to replicate those activities.

Nowadays, technology is involved in the field of the upper and lower prostheses. The type of prosthesis is selected based on what part of a limb is missing of particular importance is the above knee amputation, where the loss of the knee joint affects the movement of the whole human body. The knee joint plays an essential role in the human movement as it is responsible for carrying the body weight in horizontal (running and walking) and vertical (jumping) movements. So, technology can be used to develop a knee prosthesis that can assist the user to replicate different movements.

1.1.1 Classification of the knee prostheses

Basically, the knee prostheses are categorized into passive, active, and powered/motorized prostheses as shown in Figure 1.1. Each type of knee prosthesis has its own characteristics and functions. For instance, passive knee prosthesis consists of a mechanical structure and operating fluid that assist the prosthesis to vary the rate of damping during walking.

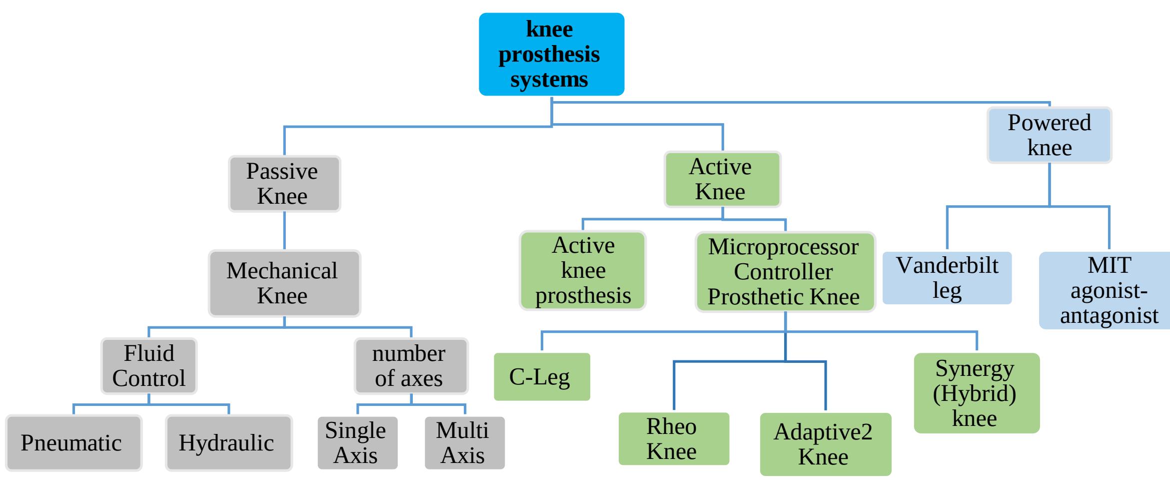

Figure 1.1: Classification of the current knee prosthetic systems

In addition, the passive knee prosthesis is classified according to its number of axes, either single axis or multi axes. A typical passive knee prosthesis behaves as a variable damper during walking. However, some weaknesses are still found in passive knee prosthesis in some movements such as stair climbing and sit-to-stand. The user requires additional torque to replicate those movements.

The evolution of the knee prostheses over the recent decades has progressed from purely mechanical systems to systems that include microprocessor, actuators, and sensors. Figure 1.2 represents the passive types of knee prosthesis, namely (a) pneumatic, (b) hydraulic, and (c) mechanical knees.

(a) (b) (c)

Figure 1.2: Various types of passive knee prosthesis devices for transfemoral amputees, (a) Heritage Polycentric Pneumatic 4 Bar Knee, (b) Heritage Single Axis Hydraulic, (Heritage Medical Equipment, IA, USA) (c) Prosthetics Freada 2SR320 Mechanical knee joint with four bar (Fujian Prosthetics Center, Fuzhou, China)

Due to the daily needs of the transfemoral amputees, passive knee prosthesis was updated to active and powered knee prosthesis as shown in Figure 1.3. Active and powered knee prostheses are able to adapt to different terrains. In addition, the powered knee prosthesis is able to generate the amount of torque and power at activities such as a stair ascent, sit-to-stand, and slope climbing.

(a) (b) (c) (d)

Figure 1.3: Various types of active/powered knee prosthesis devices for transfemoral amputees, (a) Vanderbilt leg (Sup et al., 2011), (b) MIT Biomimetic agonist-antagonist active knee (Martinez-Villalpando & Herr, 2009), (c) The Ossur Rheo knee (Ossur, CA, USA), and (d) the Ottobock C-Leg® represent the microprocessor controlled damping knee prostheses (Ottobock, TX, USA)

On the other hand, an active knee prosthesis is composed of a mechanical structure, control unit, and mechanical sensors. In particular, the function of the mechanical sensors in active knee prosthesis is to measure the knee kinematic and kinetic parameters during movement. The kinematic and kinetic parameters are the knee angle and torque, respectively. Accordingly, the sensors signals are used as a feedback to the controller to execute movement commands such as the flexion and extension of the knee prosthesis during stride. However, active knee prosthesis cannot yet assist the amputees to perform activities such as stair ascent/descent and standing up. This is because such activities require significant torque and range of motion at the knee joint and involve the coordinated movement of the entire body to substantially raise the body center of mass in a generally economical manner.

The last type of the knee prosthesis consists of a mechanical structure, control unit, actuation system, and mechanical sensors. This type is called powered/motorized knee prosthesis. The powered knee prosthesis is able to generate extra torque that is produced by the actuation system in situations such as sit-to-stand and stair ascent/descent. Also, powered knee prosthesis contains pure mechanical sensors such as angle sensors, torque sensors, and on/off switches, and those sensors are functioning independently from the motor system of the human body. As a result, the control algorithms of the knee prosthesis need some artificial intelligence, thus making it complex. In addition, the control performance of the knee prosthesis does not function as if it was a natural limb.

Sensing system is a crucial part in the development of the knee prosthesis. As previously mentioned, the knee prosthesis is composed of mechanical sensors that measures knee parameters to get kinematic and kinetic information about the prosthesis movements that should be processed by the controller. For example, a potentiometer and encoder are used in the knee prosthesis to measure the flexion and velocity of the knee. However, it is essential to use on/off switches which give information about heel strike and toe off states. Those switches are placed away from the knee prosthesis as they are attached below the prosthetic foot. For example, one of the available types of the lower powered prosthesis consists of knee and foot prostheses that are connected permanently to each other. Although the lower limb powered prosthesis has benefits of replicating walking and slope climbing movements, it is believed that both the knee and foot prosthesis cannot be used separately by the user with complete or partial amputations. The user in some situations may have to upgrade or replace the foot prosthesis with another one. As a result the user faces some difficulty to do that, because the foot prosthesis contains on/off switches that detect the transitions between phases. This concern may cause inflexibility to the user, especially if he/she needs to use a new ankle module or use a more comfortable foot or ankle prosthesis.

1.1.2 Electromyography (EMG) and electroencephalography (EEG) sensory systems in the knee prosthesis

In order to improve the interaction between the user and the sensing system of the prosthesis, attempts were conducted by multiple researchers that are related to using additional sensors. The purpose is to improve the intention of the user's movement in some daily activities. The electromyography (EMG) and electroencephalography (EEG) techniques are recently used as additional sensory system to enhance the control performance of the knee prosthesis beside the existing mechanical sensors. The EMG technique uses electrodes to detect the activity of the muscles. The output signal from the electrodes is fed to the control unit to adjust the movement of the knee prosthesis. The knee prosthesis which is controlled using EMG is named a myoelectric control of knee prosthesis, which is used by the transfemoral amputees to perform repetitive tasks in the workplace. Myoelectric control is used to control the locomotion of the knee prosthesis (Dawley et al., 2013). Some shortcomings of using EMG were observed, for example, EMG signals require amplification circuits in which developing a differential amplifier for EMG poses few real problems if the appropriate precautions are not taken. When a muscle contracts, the distribution of electrolytes within the tissue changes, which induces small voltages which need signal conditioning circuit (amplification circuit). The problem is to sense and isolate this signal so that it can be used to control the movement of a prosthesis device. In addition, electrodes that are placed in specific locations may cause discomfort and problems to the skin of the residual limb due to sweating.

Electroencephalography (EEG) is a technique that is used to measure the brain activity from the scalp (Teplan, 2002). EEG is widely used in many areas of clinical work and research. One of the biggest challenges in using EEG is the very small signal-to-noise ratio of the brain signals (Repovš, 2010). EEG signals have very small amplitudes and because of that they can be easily contaminated by noise. The noise can be electrode noise or can be generated from the body itself (Khatwani & Tiwari, 2013).

In conclusion, a part from the limitations that were mentioned earlier about EMG and EEG techniques, using EMG and EEG needs some considerations such as health and ethical procedures that may be required during the experiments. Also, a practical consideration is required when using EMG or EEG electrodes every time before and after donning and doffing the prosthesis socket. Therefore, it is recommended to search for alternative techniques that can be used as a sensing element to improve the interaction between the human body or residual limb with the control system of the knee prosthesis in order to develop the knee prosthesis to become closer to the natural limb.

1.1.3 Actuation mechanism in the knee prosthesis

The knee prosthesis is composed of an actuation system and a mechanical structure which can briefly be called the knee prosthesis mechanism. Nowadays, various types of actuators are used throughout the development of the knee prostheses. For example, magnetorheological (MR) fluid is an actuator system that uses the shear mode to produce primary torque (Herr & Wilkenfeld, 2003). MR actuator assists the controller to vary the damping of the knee prosthesis during walking. However, MR is still not able to produce sufficient torque for movements such as climbing slopes or stair ascent. Another type of actuators is electric motors that are used as an actuation system in powered knee prostheses (Sup et al., 2011). The actuation system of the knee prosthesis has to meet the daily activities of the amputees, especially movements such as sit-to-stand and slope climbing. In some situations such as sit-to-stand or stair ascent/descent, the actuation system should produce the required torque to assist the knee mechanism to rotate in a specific range of motion. In addition, the actuation system should be light in weight and compact. Bulky actuation system may cause problems and inconvenience to the amputee. The second part of knee prosthesis mechanism is the mechanical structure of the knee prosthesis which affects the overall performance of the knee prosthesis (Borjian, 2008). The design of a mechanical structure that is meant to be used in biological and human applications has to be smart, light in weight, and also durable. Moreover, the designer should take into consideration the appropriate location of the components of the knee prosthesis such as sensors, control unit, and actuation system.

During the actual situation, the actuation system may be not be able to produce continuous torque and power to move the prosthesis as expected, because of the inertia of the knee system that may need a pre-adjustment before the development process. Therefore, a physical modelling tool could simulate and control the performance of the actuation system along with the knee mechanism through the design stage. Physical simulation can mimic the real knee prosthesis mechanism at different movements. In addition, physical simulation assists the designer to update the mechanical design for better optimizing the parameters with the control system for better performance. Also, control scheme can be designed and established during the physical simulation and also tuning of the control parameters can be adjusted to investigate the appropriate control scheme. Full characteristics of the actuation system, controller, and the knee prosthesis mechanism can be obtained from the physical simulation which is useful during development stage of the real system.

The physical simulation tool also provides a feedback to the designer about the static and dynamic behaviors in terms of dimensions and mass of various knee components. The physical simulation tool is a part of the biomechatronics approach. In this thesis, this strategy has been adopted to assist in the pre-development process of the knee prosthesis mechanism.

1.2 Overview of biomechatronics philosophy

Biomechatronics philosophy is a sub-discipline of the Mechatronics approach. Biomechatronics is useful during the design and development process of a biological system or human body with mechanical, electronic, and control schemes. Basically, biomechatronics system consists of four components: biosensors, controller, mechanical sensors, and actuators (Brooker, 2012). Biomechatronics is found in biomedical applications, such as biology, medicine, health care, minimally invasive surgery, and also microgrippers that are used to grasp or cut cells and organisms, and surgical robots (Gultepe et al., 2013). The four components of the biomechatronics system are shown in Figure 1.4.

An example of the application of biomechatronics in the field of biology and human body study can be illustrated as follows. Biosensors can be used to detect the user intentions or identify the transition between states during the knee movements. In another biomechatronics system, information can be relayed by the user’s nervous system or muscle system. This information is sent by the biosensor to a controller that is located inside or outside the biomechatronics system. On the other hand, biosensors can be utilized to receive information from the position of the amputee limb, and the force generated from the limb and the actuator of an assistive device. Biosensors have variety of forms. They can be configured by wires that detect electrical activity, needle electrodes implanted in muscle, and electrode arrays with nerves growing through them. Otherwise, mechanical sensors measure information about the biomechatronics system and relate that information to the biosensor or the controller (El-Sayed et al., 2014) .

The controller relays the amputee’s intention to the actuation system. It also interprets feedback information about the user as received from the biosensors and mechanical sensors. Furthermore, the controller adjusts the performance of the biomechatronics system (Brooker, 2012).

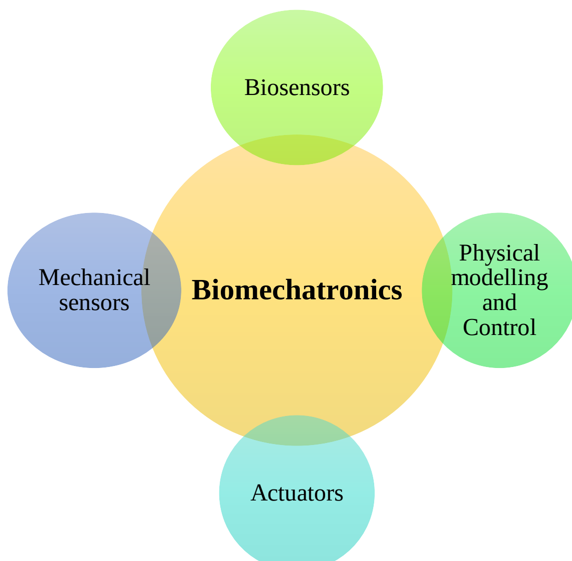

Figure 1.4: Components of bimoecharonics system include mechanical structure, sensors, actuators, and control

Biomechatronics contains measurement of physical parameters that come out either from biological system or mechatronics system. Measurement can be variables such as voltage, chemical concentration, pressure, position, and displacement. The variables are processed and then transferred into the actuation system that is responsible for moving the whole system. Therefore, selection of the sensors and actuators that are created from active components or smart materials plays a crucial role in the development of biomechatronics system with minimal complexity in the control schemes.

It is essential to get a brief information about the smart materials, or intelligent materials, that have the intrinsic and extrinsic capabilities. Firstly, they respond or stimulate environmental changes and secondly, they activate their functions according to these changes (El-Sayed et al., 2013). There are various types of smart materials, for example, piezoelectric materials are able to function as sensing and actuation elements. Once mechanical stress is applied to the piezoelectric surface, it generates an electric charge. In contrast, when the charge is applied to the piezoelectric surface, it expands and contracts according to its polarization direction. Other examples of smart materials are the magnetorheological fluid, shape memory alloys, and optical fibers. Smart materials are found in many applications such as industrial, aerospace, surgical, and medical applications (Bishop, 2007).

Nowadays, biomechatronics is related to the recent technology that is being involved in the field of assistive devices for the amputees, specifically knee prosthesis for transfemoral amputees. Knee prosthesis contains sensors, actuators, and control unit that have to function simultaneously to provide the knee prosthesis the suitable range of motion according to the type of movement. For instance, activities such as walking at different speeds could be replicated using the knee prosthesis. However, some movements such as stair ascent/descent and sit-to-stand still require sufficient continuous torque and power that should be produced by the actuation system of the knee prosthesis.

Although researchers have attempted to develop a knee prosthesis that assists the amputated person to perform various activities, movements such as sit-to-stand, stair ascent/descent, and slope climbing have still not been validated by extensive studies. In addition, the normal range of motion of the knee joint has to be achieved by the knee prosthesis. Moreover, the direct interaction between the amputee's residual limb with the knee prosthesis needs to be studied by using new sensing elements rather that EMG and EEG techniques. In particular, the control performance of the knee prosthesis can be improved if the human control system is matched with the knee controller by means of a sensing system. Researchers have to think of new techniques to improve the control performance of the knee prosthesis and, accordingly, meet the transfemoral amputees' demands.

1.3 Motivation for the study

In analyzing the existing active and power knee prostheses, it was found that knee prostheses are controlled by microprocessors that are based on artificial intelligence or pre-programmed algorithm to predict a response to environmental situations. The environment is sensed using pure mechanical sensors such as load cells and gyroscopes.

The existing knee prostheses could replicate walking. A few of them could assist in performing sit-to-stand and slope climbing movements. However, transition between states such as washing car or kicking football can be difficult to the user in some cases. This means the user's intention detection can be improved to better recognize those transitions. Also, the current knee prosthesis operates independently from the user's intention, although the user can improve the control performance of the system by moving or loading the prosthesis.

Detecting the transition between phases is a step to study the intention of the user, which is beneficial beside the pure mechanical sensors to achieve movement similar to the natural limb. EMG technique has some limitations of low voltage levels which vary from 50 µV up to 5 mV, and needs some kind of amplification circuits. Also, the contact between the electrode and the skin could distort a recording signal. Besides, EMG technique causes skin problems due to sweating, and the signal may vary due to the skin impedance. Although there are attempts to study and use EEG technique in the field of controlling prostheses, it is becoming limited because of the short life span and robustness, and noise produced by electrodes or the body.

The available knee prosthesis mechanism (actuation system and mechanical structure) has some shortcomings to replicate the basic daily movements at a normal range as well as different speeds. Moreover, the knee prosthesis mechanism has limitation to provide the normal range of motion up to 120°. The range of movement that is produced by the current prosthesis is not sufficient to assist the transfemoral amputees to imitate some activities within the normal range from 0°-120°. Thus, the current thesis presents three main questions and a few sub-questions that have to be addressed according the literature related to the knee prosthesis (Figure 1.5).

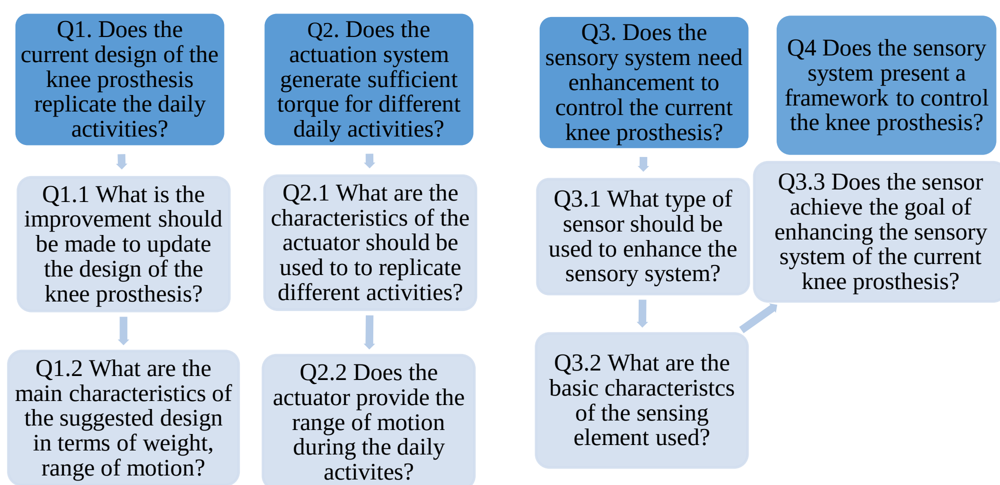

Figure 1.5: The main questions and sub-questions covered during the current study

The outcomes of the questions Q1, Q2, Q3, and Q4 require studying the current knee prostheses, in terms of the mechanical structure, actuation system, and other sensor's characteristics. Thus, a comprehensive survey was necessary to study the main components of the knee prosthesis. Accordingly, sub-questions have to be answered based on the knowledge and answers of the three main questions. Finally, the answer to question number 4 that is considered one of the main contributions of the current study, will be presented in the coming chapters.

1.4 Aim and scope of the study

The motivation for the study is the improvement in the sensory system of the existing knee prosthesis in order to better control its performance. In addition, the aim is to establish a new design of a knee prosthesis mechanism that is able to achieve anthropomorphic range of motion. The current knee prosthesis mechanism has limitation in achieving the normal range of motion as well as in emulating the different daily activities at different time intervals.

The aim specifically is to examine the possibility of using an alternative sensory system that could directly interact with motor control system of the human body. In other words, the amputee’s residual can be in direct contact with the sensory system to provide information about the intention and transition between states at different daily movements. The sensory system basically consists of sensing element that is used as a feedback to the control unit of the knee prosthesis.

Furthermore, the designing, testing, and investigating of a knee prosthesis mechanism that could function within the normal range of about 120°, and could mimic basic activities namely walking, sit-to-stand, stair ascent/descent, and slope climbing at different speeds. It is believed that the main aim of the dissertation is as mentioned in the previous paragraphs. However specific aims can be listed based on the main aim:

i. To test a new sensing element that is appropriate to be used in the field of the knee prosthesis, as well as, to work out the full characteristics of the smart piezoelectric bimorph element that has the benefits of self-sensing, without the need of external power supply that is normally required by conventional sensors such as inductive, resistive, and conductive types.

ii. To simulate and test the knee joint mechanism for basic daily movements, to make sure that it provides sufficient torque and power, as well as, to adjust parameters such as mass and inertia of the knee mechanism.

iii. To check the normal range of motion for most of the daily activities which varies from 0°-120°, to set the required parameters of the knee prosthesis mechanism to replicate basic daily movements, and to establish a framework of controlling the knee prosthesis via an in-socket sensor, where the framework presents the integration between the actuation system, in-socket sensor, and control unit. Moreover, the framework proposed works as a platform for the future research in the field of the lower prosthesis.

1.5 Structure of the thesis

The thesis is organized into seven chapters, as follows:

Chapter 1 presents the introduction, motivation, research questions, and scope of the thesis.

Chapter 2 reviews the current technologies of the knee prostheses in terms of sensors, actuators, and control methods.

Chapter 3:

Paper 1: Piezoelectric Bimorphs' Characteristics as in-Socket Sensors for Transfemoral Amputees.

"This study presents the use of piezoelectric bimorphs as in-socket sensors for transfemoral amputees. Static and dynamic characteristics of the piezoelectric bimorph were investigated. This paper highlighted the capacity of piezoelectric bimorphs to be functioned as in-socket sensors for transfemoral amputees".

Paper 2: Development of a Micro-Gripper Using Piezoelectric Bimorphs.

"In this study, the dynamic behaviors of bending piezoelectric bimorphs actuator were theoretically and experimentally investigated for micro-gripping applications in terms of its deflection along the length, transient response, and frequency response with varying driving voltages and driving signals. In addition, its implementation as a parallel micro-gripper using bending piezoelectric bimorphs was presented. The micro-gripper could perform precise micro-manipulation tasks and could handle objects such as a small strain gauge".

Paper 3: Detection of Prosthetic Knee Movement Phases via in-Socket Sensors:

"This paper presents an approach of identifying the movements of the knee prosthesis through pattern recognition of mechanical responses at the internal socket's wall. Force sensitive resistor (FSR) and piezoelectric outputs were measured with reference to the knee angle during each phase. Piezoelectric sensors could identify the movement of midswing and terminal swing, pre-full standing, pull-up at gait, sit-to-stand, and stair ascent. In contrast, FSR could estimate the gait cycle stance and swing phases and identifying the pre-full standing at sit-to-stand. The study highlighted the efficacy of using in-socket sensors for knee movement identification".

Paper 4: Control the performance of a linear actuated transfemoral knee joint in basic daily movements.

"This paper presents new proposed design of an actuated knee prosthesis mechanism that consists of an actuation system that is capable of feeding the knee's mechanism with the required moment and power at different movements. The PID control parameters were tuned until the measured angle of the actuated knee mechanism could track the desired angle within a time period of 1s and 0.1 s, however, the mechanism shows the deviation from the input at time periods of 0.05 s and 0.0125 s".

Chapter 4 presents the outcome of the research questions and discusses the correlation between the obtained results with the existing studies.

Chapter 5 provides the conclusion and recommendations for future studies.

A diagrammatic view of the thesis structure and the four chapters is shown in Figure 1.6. The diagram shows the storyline of the current thesis, including all the thesis chapters and their relation with respect to each part of the thesis.

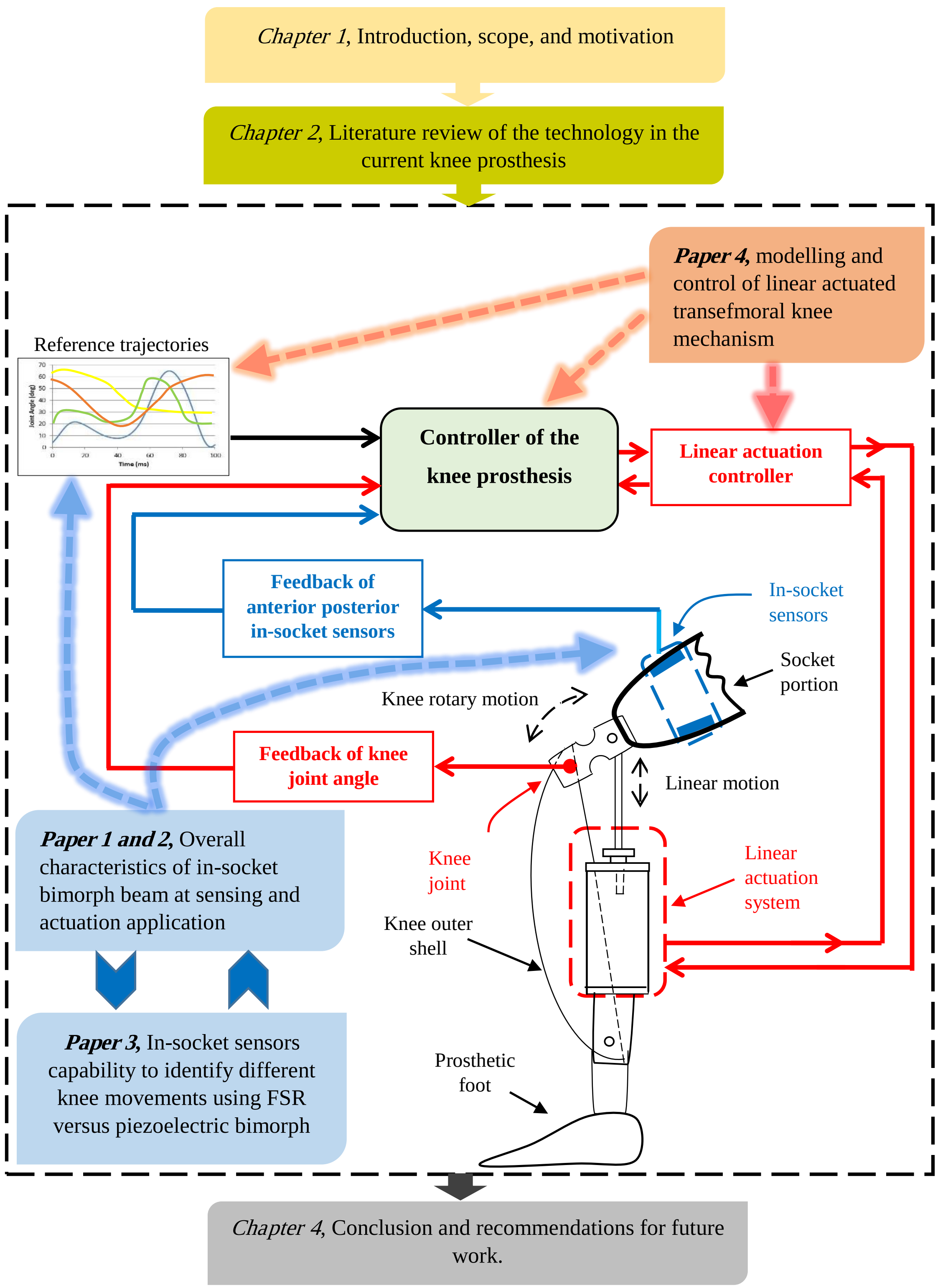

Figure 1.6: A diagram concludes the thesis chapters and their relation to the thesis sections

CHAPTER 2 : LITERATURE REVIEW

2.1 Introduction

A knee prosthesis is the key component of a transfemoral prosthesis prescribed to an above knee amputee as it replaces the functional joint. The development of knee prosthesis has increasingly enabled amputees to perform activities of daily living (ADL). The variation in the needs of daily movements of the persons who lost their lower limbs has encouraged scientists to continuously improve the sensory system and the knee prosthesis mechanism.

This chapter shows the recent advancement of technology in the knee prosthesis. Different types of knee prostheses are classified according to their functional mechanism. Also, sensory system, actuation system, and control system in the knee prosthesis are presented. Moreover, weight consideration of the current knee prosthesis and the power supply sources are discussed. Furthermore, current types of prosthetic foot that may affect the biomechanics of the knee prosthesis are also presented.

2.2 Passive knee prosthesis

Passive knee prosthesis. has fixed impedance that is provided from either pneumatic or hydraulic systems, in which the friction of the prosthesis might change according to the walking speed of the amputees (Radcliffe, 1977). Unlike active prosthesis, passive knee types are not equipped with sensors that aid the interaction between the amputee and the environment (Alzaydi et al., 2011; Dunmin et al., 2011). For example, Mechanical–based prostheses employ passive prosthetic knee systems that have limited ability to mimic the behavior of a normal knee (Jay Martin, 2010). In a manual locking knee, a remote release cable is utilized to provide the user stability during knee extension. However, this device leads to high energy costs during ambulation. In weight-activated knees, constant-friction is used to supply high stability during stance phase. Transferring the body weight to the knee shall activate an embedded brake that prevents buckling. This brake is released once the knee is unloaded. However, constant friction is still presents during the swing phase and these results are inefficient during gait. An element that is capable of storing energy, such as a spring may assist the knee during the swing phase in which it is loaded during weight bearing and released during the swing phase itself (Martinez-Villalpando & Herr, 2009; Michael, 1999). Another type of passive knee device is the single axis knee, which utilizes a simple pivot mechanism. It is robust, lightweight, and relatively cheaper than other knee systems that are commonly in use. However, amputees have to use their own muscles to maintain the device stable at standing, as it is not equipped with a stance control function. Thus, users often use manual lock to compensate for lack of stance control and utilize the available friction to prevent the leg from over-speeding during the forward swing when moving into the next step. Polycentric knees offer an advantage in terms of low maintenance but do not contribute towards walking pattern that resembles an able-bodied person. Other than the spring-loaded friction, polycentric knees offer little or no stance control. Stance control is a very important feature that prevents the knee from buckling in the event of an accident or an unexpected change during gait control. Nevertheless, some polycentric knees incorporate a simple locking mechanism that allows the knee to be locked in the extended position (Michael, 1999). Such passive knee devices cannot adjust or control the amount of power during different terrains.

2.3 Powered/ motorized knee prosthesis

Another type of knee prosthesis is the powered knee prosthesis, that has the capability to assist amputees to perform level walking, ramp descent, stair descent, standing, and detect instances of stumble (Wang et al., 2007). However, the devices still has some limitation to deliver sufficient joint power to restore functions such as climbing stairs, running, many locomotive functions, and standing stability during slope terrain (Dedic & Dindo, 2011; Lawson et al., 2011; Song et al., 2008). On the other hand, powered knee devices can reduce the hip power demand during stance and swing phase (Hoover et al., 2013). External power delivered to the prosthetic knee enables adaptation at different walking environments. In addition, different types of sensors can be utilized to provide interaction between the prosthetic knee and the external environment. To date, there are two models of the powered knee devices: Ossur power knee® and the Vanderbilt leg (Sup et al., 2011). The two models have motors that generate power to facilitate the user to perform (ADL).

To improve the mobility of the knee prosthesis system, it incorporated the ability to restore user stability. The knee prosthesis was referred to as active device, as it is capable of delivering active response to prevent the amputee from falling due to obstacles. One example of those active powered knee can be seen in the stumble detection feature in which real time stumble can be detected using three separate accelerometers on the prosthesis. Ten subjects were employed to validate the stumble detection through these signals and all 19 stumble responses were correctly identified (Lawson et al., 2010). Further study was performed to enhance the interaction between the user and the environment whereby new knowledge of adaptive locomotion-mode-recognition system was developed to enhance the performance of the prior system. That system was based on the integration of EMG and mechanical signals, which showed great potential at locomotion-mode recognition (Du et al., 2012). EMG based control was successfully used to control the active knee prosthesis as presented through a set of swing experiments (Wu et al., 2011).

2.4 Adaptive dissipative knee prosthesis

Adaptive dissipation knees refer to devices that can regulate the impedance by adjusting the hydraulic valves such as the Ottobock C-Leg® or the devices that use magnetorheological fluid such as Rheo knee™. Microcontroller on board can regulate the impedance by tuning hydraulic valve, i.e. the orifice that rectifies the flow rate. Adaptive knees adjust the resistance to control the knee after calculating the comparisons between steps and monitoring the position, velocity, forces, and moments of the prosthetic knee (Hoover et al., 2013). To differentiate between passive and active knees in terms of the advantages of each type to the transfemoral amputees, a clinical comparison was performed to identify the damping in mechanical passive and active prosthetic knee devices. The study established the idea that variable-damping knee prostheses offer transfemoral amputees significant advantages over mechanically passive designs (Johansson et al., 2005). To solve the problems associated with passive knee systems and in order to enhance amputees' functional performance, researchers started to involve electronics in the design of prosthetic knees. For example, a prosthesis that consisted of a hydraulic actuator tethered to an external source power was used to move the knee joint and subsequently control the knee (Stein & Flowers, 1987).

In general, the working principle of a smart or active device involves the integration of intrinsic computational sensors, whereby the sensors detects the physical performance of the system and this corresponds to real-time alterations in the actuator. However, an embedded controller adjusts and coordinates the knee movement accordingly (Jay Martin, 2010). The first version of an active prosthetic has been developed and the electronic prosthesis is now available as a product on the wider market (Popovic & Schwirtlich, 1988). The Ottobock C-Leg®, Össur Rheo Knee®, Adaptive2®, and Synergy knee (Bellmann et al., 2012) are all active adaptive dissipation knee. Nevertheless, the prior knee systems are still unable to generate sufficient mechanical power for normal ADLs. The experiences of nine transfemoral amputees who wore C-Leg®, Rheo knee®, and Adaptive2®, were acquired in order to identify which of these commercial knee systems offer functional benefits to amputee. The results revealed that C-Leg® could offer the most functional benefits during everyday gait (Bellmann et al., 2010).

One study compared the kinetics and kinematics performance of the C-Leg® and the Mauch SNS® knee prosthesis during gait (Segal et al., 2006). The study reported that the C-Leg® offers low performance in terms of knee angle, knee moment, and knee power with comparison to the Mauch SNS®. A review study presented transfemoral amputees' opinion of the C-Leg® microprocessor-controlled prosthetic knee in terms of three categories: safety, gait energy efficiency, and cost effectiveness. The study concluded that C-Leg® is safe, energy efficient, and cost effective compared to other prosthetic knee systems (Highsmith et al., 2010).

On the other hand, it is important to identify and evaluate the performance of each sensory, actuation and control element of knee prosthesis, a more crucial element in designing the knee prosthesis is to build a knee mechanism that has the capability of performing functional tasks such as sit-to-stand and stair climbing. Researches nowadays are investigating better prosthetic devices to assist the amputees during sit to stand and stair climbing. The functional capability that requires modulation of the power output of the limb should be achieved with the knee joint to contain the motorized actuators, sensors, and the controller. The following sections discuss the components of various technology utilized in the retrieved studies.

2.5 Sensors in the knee prosthesis

A prosthetic knee system requires many input variables to perform optimally, thus different types of sensors are used in the knee prosthesis. To select appropriate sensory systems for the knee devices, the control variables have to be determined. Adaptive-control of prosthetic knee (Herr & Wilkenfeld, 2003), was reported to monitor two types of variables namely, force/moment and flexion parameter. Microprocessor acquires information from the sensors to vary the resistance of the knee movement during gait. Strain gauges were located at appropriate positions close to the knee axis to identify a suitable correlation of the knee joint movement. Angle sensors were attached at the knee axis to measure the knee extension/flexion during walking. The sensors were used to duplicate the normal gait cycle for the adaptive-control prosthetic knee by varying the impedance. Active artificial knee joint (Kapti & Yucenur, 2006), provides the tracking position of the knee joint during the gait cycle. Measuring the knee flexion and identifying the appropriate knee position during gait cycle can assist tracking of the knee position reference trajectory. Angle rotary sensor was attached to be able to perform that task. Agonist-antagonist knee prosthesis (Martinez-Villalpando & Herr, 2009; Martinez-Villalpando et al., 2008), used knee angle sensor, hall effect sensor, and force sensitive resistor (FSR). Different impedance was identified by the controller during the stance and swing phases. All the described technology related to the mentioned researches employed different sensory systems integrated with the knee devices based on the controlled parameters.

Both angle sensor and moment or torque sensors plays a prominent role in developing an active knee. In order to record heel strike and toe-off to provide information about walking phases, FSR is used, located directly below the prosthetic foot. Furthermore, it is low in price, relatively thin, small, and produces analog based signal (Syrseloudis et al., 2008). Basically, the FSR sensor is a pressure sensor that is used to detect the walking phase of the prosthetic knee. FSR provides sufficient information about different walking phases, but as the FSR is located below the prosthetic foot itself, replacement of the foot would require technical adjus^{tm}ent, which may be quite impractical amongst normal prosthetists or amputee users themselves. In order to improve the control performance of the knee prosthesis, sensors should be located near to the knee axis. Also, the direct interaction between the residual limb and sensor will improve the user intention as well as the control of the knee. However, most of the configurations have limitation further to the interaction between user and knee device, especially in detecting transition between phases and the user intention. Some studies presented EMG and electroencephalography (EEG) that are used to improve the intention and control of the knee prosthesis.

The detection of muscle activity can be detected using electromyography (EMG). The EMG technique is used to measure the muscle activities of the residual limb that is consequently fed to the control unit to control the performance of the knee prosthesis. The knee prosthesis that is controlled using EMG is named a myoelectric control of knee prosthesis, and these were being used by the transfemoral amputees to perform repetitive tasks in the workplace. Myoelectric control was used to control the locomotion of the knee prosthesis (Dawley et al., 2013). Some shortcomings of using EMG were observed, the output amplitude of the EMG signals is measured in mV, which requires additional signal conditioning circuits to process them. In addition, electrodes that are placed in specific locations may cause discomfort and problems to the skin of the residual limb due to sweating.

Electroencephalography (EEG) is a technique that measures the brain activity from the scalp (Teplan, 2002). Although there are attempts to study and use EEG technique in the field of controlling prosthesis, it is becoming limited because of the short life span and robustness

(Hardaker et al., 2014). On the other hand, using EMG or EEG techniques needs some considerations such as health and ethical procedures that are be required during the experiments. Therefore, it is recommended to search for alternative techniques that can be used in the development of the knee prosthesis for better performance.

2.6 Actuators in the knee prosthesis

The natural knee produces an internal flexor moment, due to contraction of the hamstrings, which prevents hyperextension at the end of the swing phase. As the knee starts to flex, concentric contraction of the hamstrings, as well as the release of energy stored in the ligaments of the extended knee, results in short-lived power generation (Hollander et al., 2005). For amputees who have lost their lower limb, utilizing an appropriate actuation system could facilitate walking and other daily activities. Actuators used in damping strategies (Herr & Wilkenfeld, 2003), can be categorized as: hydraulic, pneumatic, and magnetorheological (MR). However, motorized actuation was used in knee prosthesis technology to deliver positive power to assist the amputee in performing some ADLs such as generating sufficient power during stair ascent/descent, stand/sit, as well as slope climbing. Motors are connected to gear box and lead screw assembly to generate the required moment and knee power. For example, electric motors are used as an actuator in powered knee prosthesis. In addition, the actuation system should be light in weight and compact. Otherwise, a bulky actuation system may cause problems and inconvenience to the amputee. It is expected that the actuation system will produce the required mechanical torque and power during the actual movement of the knee prosthesis. In contrast, the actuation system during the actual situation may be not be able to produce continuous torque and power as expected, because of the inertia of the system that may need a pre-adjustment.

2.7 Control scheme in the knee prosthesis

Control strategy is considered as the critical part in a knee prosthesis. Impedance control algorithm was the most commonly used control strategy, in which the torque generated is adapted to the produced knee angle. This mode of control ensures that the knee joint generates torque that is suitable for each gait phases (Martinez-Villalpando & Herr, 2009; Sup et al., 2007; Sup et al., 2008). Another type of control algorithm is the tracking control, whereby the joint was made to follow or track a pre-defined trajectory (Geng et al., 2010). This tracking method controls angle and velocity of the knee joint during stance and swing phase. The control strategy is related to the sensors that are involved in the knee prosthesis. the control strategy can be improved if the sensors are located in direct contact with the human body or the residual limb, some attempts are found to detect the intention and transitions between phases in order to improve the whole performance of the knee prosthesis.

2.8 Weight Considerations in the knee prosthesis

Weight of the prosthesis is an important aspect during the development of the knee prosthesis. Nowadays, researchers try to optimize the weight during the developing the prosthetic knee. In addition, accompanying devices and weight reduction, strength and functionality are usually the primary goals in prosthetic fabrication. Metal is mainly used for rotating components whereas aluminum as an alternative to steel is used conservatively for smaller components. Titanium is more expensive, however due to its biocompatibility, it is considered for some aspects of prosthetic devices (Bradley, 2010).

Lower limb prostheses need to support the body weight during the stance phase and must prevent the knee from sudden joint flexion (Herr & Wilkenfeld, 2003). Lighter prosthetic knee may also provide more comfort to the user when walks on sloped terrains or up and down stairs. The overall weight of the active prosthesis can be reduced by minimizing the overall volume of the selected actuators (Sup et al., 2007). Knee prosthesis should not exceed the size and weight of the missing limb, thus the weight of the transfemoral prosthesis plays an essential role in its development. Table 2.1 classified the corresponding weight of the current knee prosthesis.

Table 2.1: Weight of the current knee prosthesis

Author	System description, volunteer's weight, and walking speed	Weight
Ernesto et al., 2009	Volunteer mass= 97 kg, walking speed = 0.8 m/s.	3 kg.
Kapti and Yucenur, 2006	Titanium, aluminum and polyamide materials were used. Volunteer mass and walking speed were not cited.	6.9 kg.
Sup et al., 2007	A tethered transfemoral prosthesis with pyramid connectors. Volunteer mass= 75 kg, walking speed= 0.7 m/s.	2.6 kg.
Sup et al., 2008	Volunteer mass= 85 kg, walking speed= slow, normal and fast (2.2, 2.8 and 3.4 km/hr) respectively.	3.8 kg.
Fite et al., 2007	Volunteer mass= 85 kg, walking speed= 0.8 m/s.	3 kg.
Torrealba et al., 2010	Volunteer mass (not cited), walking speed= self-selected speed.	2 kg.
Sup et al., 2011	Volunteer mass= 80 kg, walking speed= 5.1 km/h.	4.2 kg.
Gong et al., 2010	Volunteer mass= 62 kg, walking speed= slow, normal, and fast (0.7, 0.7, and 1.2 m/s) respectively.	Not cited.
Geng et al., 2010	Volunteer mass= 62 kg, walking speed= slow, normal, and fast of average values (1, 1.2, and 1.5 m/s) respectively.	Not cited.
Hoover et al., 2013	Volunteer mass= 83 kg.	3.5 kg.

2.9 Operation and power sources in the knee prosthesis

The sensors and actuators that are incorporated in the knee prosthesis normally require electrical power source for their operation. Different power sources of the existing prosthesis are listed in the power sources that mentioned in the available studies are listed in Table 2.2. New techniques of providing alternative power source from smart material such as piezoelectric are adopted by the researchers. Harvesting power circuit that can be build using piezoelectric bimorph is capable to deliver about 5.2 V. The smart piezoelectric material can be placed below the foot to charge some amount of power that can be saved and used to operate electronics circuits and controller. These techniques might decrease the weight of the battery while saving some power and can also be utilized as an emergency power source (Almouahed et al., 2011; OBE et al., 2005).

Table 2.2: Power sources in the prosthetic knee systems

Author	Power Source for various prosthetic knee systems	Weight
Ernesto et al., 2009	A 6-cell Lithium polymer battery with 22.2 V nominal rating.	0.2 kg.
Gong et al., 2010	Rechargeable lithium ion battery.	Not cited.
Sup et al., 2011	A lithium polymer battery with 29.6 V nominal rating and 4000 mAh capacity.	0.8 kg.
Fite et al., 2007	A high-power Li-ion battery with nominal capacity of a single battery is 2.3 Ah and 3.3 V.	0.1 kg.
Hoover et al., 2013	Four 11.1 V, 2000 mAh lithium polymer batteries.	0.1 kg.

2.10 Overview of prosthetic foot type

Prosthetic foot is considered one of the lower prosthesis components, which may have an effect on the biomechanical outcomes of the knee prosthesis (Van der Linde et al., 2004). Therefore, different types of prosthetic foot that are incorporated in the lower prosthesis systems are presented in Table 2.3. Prosthetic foot affect the selection of the suitable control scheme of the lower prosthesis at different locomotion such as ground-level walking, sitting/

standing mode, and stair ascent/descent (Jimenez-Fabian & Verlinden, 2012). In overall, most systems used conventional passive prosthetic foot except (Sup et al., 2011), which contains a custom sensorized foot.

Table 2.3: Prosthetic foot devices in the prosthetic knee systems

Author	Prosthetic foot type
Ernesto et al., 2009	Conventional passive-elastic ankle-foot prosthesis, flex-foot, VariFlex from Össur®.
Kapti and Yucenur, 2006	No specific type cited.
Gong et al., 2010	No specific type cited.
Sup et al., 2011	Custom Sensorized prosthetic foot.
Hoover et al., 2013	Commercial low-profile prosthetic foot, Lo Rider from Ottobock®.
Sup et al., 2008	Custom Sensorized prosthetic foot.
Fite et al., 2007	Low profile prosthetic foot, Lo Rider from Ottobock®.
Torrealba et al., 2010	No specific type cited.

2.11 Summary

This chapter presented a brief review regarding the existing technology in the knee prosthesis. It is noted that, in order to achieve a more realistic prosthetic knee system for transfemoral amputees, additional sensors might be integrated in the knee prosthesis to recognize the transition of movement for the amputee, so alternative sensing element may overcome the limitations of the EMG and EEG techniques. From a biomechatronics point of view, the integration of different types of sensors and actuators is possible to enhance the performance of the whole knee prosthesis The main challenge in the development of knee prosthesis is the ability of the sensory system to identify precisely the transition between different phases at various daily activities. On the other hand, the knee prosthesis mechanism has to be able to replicate the normal range of motion of various activities. Furthermore, the actuation system should produce the required torque and power at different daily movements.

CHAPTER 3

Paper 1: Amr M. El-Sayed, Nur Azah Hamzaid, Noor Azuan Abu Osman. Piezoelectric bimorphs' Characteristics as in-Socket sensors for transfemoral amputees. *Sensors,* 2014, 14(12), 23724-23741.

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Article

Piezoelectric Bimorphs' Characteristics as In-Socket Sensors for Transfemoral Amputees

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Abstract: Alternative sensory systems for the development of prosthetic knees are being increasingly highlighted nowadays, due to the rapid advancements in the field of lower limb prosthetics. This study presents the use of piezoelectric bimorphs as in-socket sensors for transfemoral amputees. An Instron machine was used in the calibration procedure and the corresponding output data were further analyzed to determine the static and dynamic characteristics of the piezoelectric bimorph. The piezoelectric bimorph showed appropriate static operating range, repeatability, hysteresis, and frequency response for application in lower prosthesis, with a force range of 0–100 N. To further validate this finding, an experiment was conducted with a single transfemoral amputee subject to measure the stump/socket pressure using the piezoelectric bimorph embedded inside the socket. The results showed that a maximum interface pressure of about 27 kPa occurred at the anterior proximal site compared to the anterior distal and posterior sites, consistent with values published in other studies. This paper highlighted the capacity of piezoelectric bimorphs to perform as in-socket sensors for transfemoral amputees. However, further experiments are recommended to be conducted with different amputees with different socket types.

Keywords: in-socket sensor; piezoelectric bimorph; stump/socket pressure

1. Introduction

Advancements in prosthetic knee systems are of increasing importance to assist transfemoral amputees perform their different daily activities [1] such as walking, stair climbing, and running [2,3] more naturally. Prosthetic knee devices are categorized into passive and active types [4]. In order to assist the amputees to replicate such daily movements, active knee devices have to be used to perform those functions. Active knee systems imply that the amputee can interact with the device to facilitate his/her movements. In other words, improving the sensory system of the active knee device shall assist amputees to perform their activities better and more efficiently. Therefore, the development of a prosthetic knee control system is related to sensory signals which facilitate the design of the control algorithm [3,5]. Different types of sensors are involved in active knee devices, for example, a potentiometer acts as an angle sensor to measure the knee joint angle, a load cell is used to measure the knee torque, a gyroscope sensor to detect the acceleration of the knee joint, and a force sensing resistor (FSR) is utilized as on/off sensor to detect the prosthetic knee phases [6]. Each sensor measures a certain parameter. For example the angle sensor (potentiometer) measures the inclination angle of the knee joint during the stride, while a torque sensor identifies the amount of torque that is needed for the knee to perform the movement [7]. These sensors are called passive sensors [6,8,9], as they are placed around the prosthetic knee joint to identify the knee movement. Nevertheless, the interaction between the socket and the amputee subjects is not involved in identifying the knee movement. The direct contact between the amputee subject and the socket device in the presence of the in-socket sensor would be more useful to acquire direct measurements from specific socket locations.

So far, to receive input signal from the stump muscles, electromyography systems (EMGs) were used to detect the muscle activities. An EMG embedded in an active knee system reads the interaction from the user as they detect the user's flexor and extensor muscle activities, generally from the rectus femoris, vastus lateralis, vastus medialis, biceps femoris, and semitendinosus. In order to make use of the EMG signals, such signals are analyzed to formulate the control algorithm that assists the amputee to control the torque in activities such as stair ascent/descent. However, EMG signals measure the muscle activity without considering the reaction forces and moments generated from the ground via the socket by means of pressure distribution. Considering the measurement of the pressure distribution inside the socket that originated from the ground reaction forces to understand the stress distribution during stride might be useful for gait phase identification.

Unlike measuring static pressure distribution such as the interface pressure on the buttocks, the pressure characteristics between the prosthetic socket wall and the stump would have a dynamic interaction between the socket interface pressure [10]. One example of transducer is the load cell, which has different types such as strain gauges to detect force in various applications [11]. Strain gauges are also being used in applications such as wind-tunnel balances and force sensors for robot linkages [12]. Measurement of the interactive forces between human hand and limb rehabilitation devices is achieved using a custom four degree of freedom strain gauge [13]. However, strain gauges show better behavior for static force measurements rather than dynamic investigations, as they show some limitations in the transient responses compared to piezoelectric (PVDF) materials [14]. Another sensing element that is appropriate for detection of dynamic measurements is the piezoelectric material. A piezoelectric bimorph is considered an active element, thus no external power is required to activate the sensor [15]. Moreover, one advantages of the piezoelectric bimorph is that it can adapt to vibrations in such dynamic applications. Piezoelectric bimorphs are among the most widely used sensors in academic research and industrial applications [16].

Piezoelectric materials with a bimorph configuration are used as sensors/actuators in many fields including industrial, aerospace, and medical systems [17–19]. Upon applying a load to the surface of the bimorph, an electrical charge is produced. The relation between the applied force *versus* the piezoelectric bimorph and the output deflection is essential in surgical applications and micro-gripping of fragile objects [16,19]. The charge generated inside the bimorph is measurable in volts, which is proportional to the load applied across its surface [16]. Because of the ability of the piezoelectric bimorph to be used as both the sensor/actuator element [19], current research aims to leverage the advantage of using the piezoelectric bimorph as a sensing element to detect the distribution of the pressure in transfemoral amputees' stump/sockets. In addition, the approach presented in this paper should provide better understanding of the gait characteristics of transfemoral amputees and assist the fabrication process of various socket types [20]. The appropriate location of the sensor inside the socket's wall would provide flexibility to the amputee while wearing the socket and improve the interaction during different activities. On the other hand, researchers in the lower prosthesis field are searching for alternative techniques to improve the sensory system of prosthetic knee devices. Such techniques shall assist the amputee to interact with his/her prosthesis via the sensory system. Thus, sensory system selection may assist the implementation of the control algorithm of the prosthetic knee and could provide alternative solutions for measurement of the interface pressure inside the stump.

Another challenge nowadays is how to find new methods of measuring the interface pressure for transfemoral amputees. Measuring the interface forces between the socket and the stump could provide information about the socket fabrication in the lower amputation field [21,22]. To date, the interface pressure for transfemoral amputees has not been clearly investigated, due to the shape of the stump that may vary from one amputee to another. Researchers have attempted to predict the amount of forces generated inside the stump of transtibial and transfemoral amputees. One study on interface pressure inside the stump measured it for transtibial amputees using F-socket transducers 9811E (Tekscan, Inc., South Boston, MA, USA) in which the transducers were attached to the posterior, anterior, lateral, and medial compartments of the stump to obtain better insights into the pressure between the stump and socket. The trials were conducted for the amputees during stair ascent and descent, and the study revealed that a high interface pressure exists between the stump and socket with the Seal-In X5 interface system [23]. A Flexforce network sensor made up of five Flexforce elements was used to measure the pressure inside the stump for transfemoral amputees, in which the study reported the amount of forces that can be measured at the x-direction which was about 26 N [24]. Another attempt was performed by using a Fiber Bragg grating (FBG) sensor that was developed to measure the interface pressure between stump and interface socket for transtibial amputees [25], where the range of measurement of the FBG was reported to be about 30 N. The study reported acceptable behavior of the FBG in terms of linear relationship between the shift in the peak wavelength and the applied force. The piezoelectric bimorph can be easily embedded inside the socket to measure the interface pressures at specific regions of the lower limb where high pressure is expected, such as at the posterior, posterior distal, or interior regions [26]. In this paper an investigation of the usage of piezoelectric bimorphs in the field of prosthesis is reported. More specifically, the current approach aims to determine the static and dynamic behavior of the piezoelectric bimorphs in order to utilize them as a sensory system inside a transfemoral amputee's prosthesis socket. Moreover, transient and frequency response analysis were performed to provide information about the response time and the frequency response which would provide useful information during dynamic applications. To validate the piezoelectric bimorph performance in a real situation, an experiment with a single transfemoral amputee subject was conducted while wearing a socket embedded with piezoelectric bimorphs that were placed at different socket sites. The experiment aimed to identify the variation of piezoelectric bimorph performance at different socket sites during the amputee's stride.

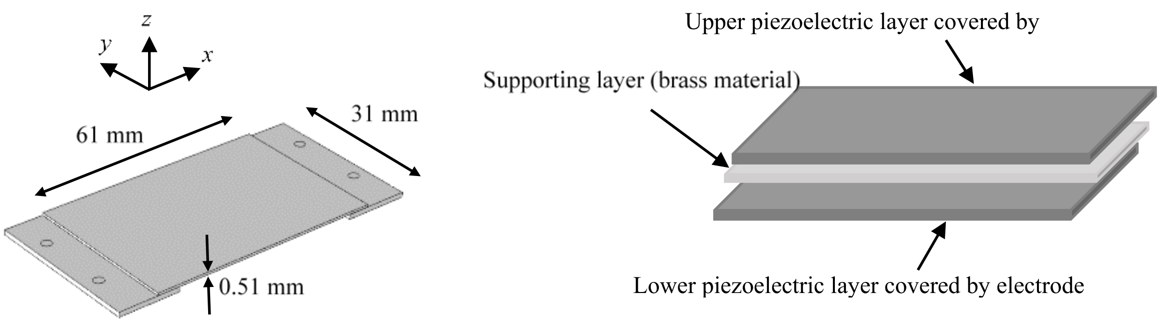
2. Materials and Methods

A piezoelectric bimorph (T220-A4-503X, Piezo Systems, Inc., Woburn, MA, USA) was selected as the sensory element in this work. Its static and dynamic characteristics were investigated. The piezoelectric bimorph was intended to be utilized as a sensing element inside the socket of transfemoral amputees. Detailed procedures for investigating the static and dynamic characteristics of the piezoelectric bimorph are presented in the following sections.

2.1. Piezoelectric Bimorph Characteristics

In order to assess the overall characteristics of the piezoelectric bimorph, a calibration procedure was conducted to estimate the static and dynamic behavior of the piezoelectric bimorph. The piezoelectric bimorph consists of two layers sandwiched by brass layer as shown in Figure 1.

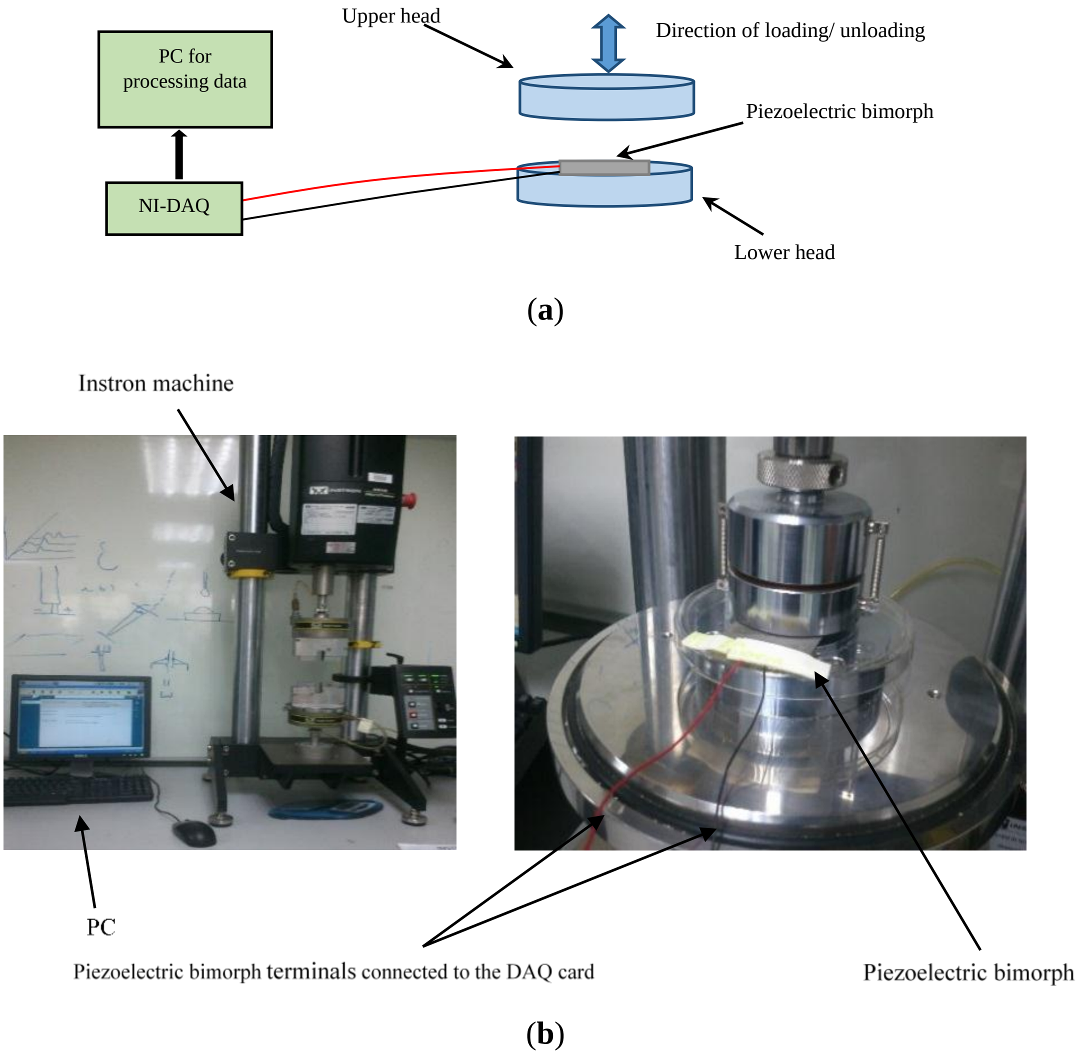
Figure 1. (**a**) Basic dimensions of the piezoelectric bimorph in a simple supported beam configuration; (**b**) Piezoelectric bimorph consists of two layers sandwiched with supporting layer.



(**a**) (**b**)

A series of input signals were applied to the input of piezoelectric bimorph and its corresponding outputs were recorded. In the current approach, static and dynamic calibrations were performed on the piezoelectric bimorph. Figure 2a shows a simple schematic of the calibration experimental setup. Also, an Instron machine (Instron Worldwide Headquarters®, Norwood, MA, USA) was used to perform the calibration of the piezoelectric bimorph as illustrated in Figure 2b. The machine consists of two lower and upper heads, while the bimorph was fixed on the lower head with the force exerted from the upper head. The purpose of the bimorph calibration is to set the static and dynamic behaviour of the bimorph that is used in development and fabrication of the active amputee's socket [21]. The calibration procedure was conducted by applying specific loads to the piezoelectric bimorph in both static and cyclic form to mimic the real situation of pressure dynamics inside the socket.

Figure 2. Overall diagram of the sensor calibration. (**a**) Simple schematic illustrating the calibration experimental setup; (**b**) Experimental setup.



2.2. Calibration Procedure

2.2.1. Static Characteristics

In this section the static characteristics of the piezoelectric bimorph were investigated. To predict the static characteristics, a series of independent-time input values were sent to the piezoelectric bimorph and the output will increase to a level that is proportional to that input. Independent-time values mean that values of inputs do not change with time. The output will remain at that level until the input level is changed. Static characteristics such as sensitivity, hysteresis, range, linearity, and repeatability were referenced in the current work to evaluate the piezoelectric bimorph's response. The benefit of the calibration is to get the characteristics of the sensor, one of the calibration techniques that can provide accurate measurements and collect data in short time is the motion and shape approach [12]. However, one of the limitations of that technique appear in the dynamic measurements as the sensor should be moved with minimum acceleration to make the measurements quasistaic. In addition, the common calibration method of the force sensor is performed by loading input forces on the sensor element. Afterwards the output voltage is recorded [27,28]. The common calibration procedure uses a loading device such as a loading plate, weights, and such a base. Here, the calibration technique that was adopted uses a standard calibration machine in which, a known force value from an Instron (Microtester 5848) strain machine was produced. The corresponding voltage output from the bimorph was recorded simultaneously. The calibration was done in the range of interest, because measurements within the range of interest will assist to enhance the bimorph's sensitivity and resolution. A schematic view of the piezoelectric bimorph that was placed between the machine's heads is shown in Figure 2b. The applied compressive force started at 0 N and increased up to 100 N. The rated deflection and force were recorded *versus* the corresponding output voltage of the piezoelectric bimorph. Data were acquired with both increasing and decreasing loads steps to highlight the hysteresis characteristics. The sensitivity of the piezoelectric bimorph was determined by calculating the slope of the static calibration curve.

2.2.2. Dynamic Characteristics

One of the significant characteristics of the bimorph is the capability to measure different parameters such as force or displacement while the input varies with time. Dynamic characteristics show the behavior of the bimorph during dynamic applications. Each bimorph has the ability to measure static and dynamic movements up to a specific range. Basically, the piezoelectric bimorph was evaluated to predict its behavior when exposed to a family of variable dynamic input waveforms such as a sinusoidal function to obtain the frequency response and a square signal to find out the response time and the damping [29].

The transfer function can be derived to attain a relation between input and output of the piezoelectric bimorph. The piezoelectric bimorph element can be modeled as a simple vibratory system (spring-mass-damper system) [30,31], that presents the analytical dynamic behavior of the piezoelectric. The output voltage *versus* the input force can be provided in terms of damping coefficient and frequency. The dynamic response of the bimorph was described as a second-order system a Laplace form as in Equation (1) [24]:

$$\frac{C(s)}{R(s)} = \frac{\omega_n^2}{s^2 + 2\xi\omega_n s + \omega_n^2} \tag{1}$$

where $C(s)$, the output of the system, $R(s)$, input to the system, ω_n, natural frequency of the system, ξ, Damping of the system.

The behavior of the second order system is described by ξ and ω_n; as an assumption damping of $\xi = 1$ is considered. Therefore, $C(s)$ for $R(s) = 1/s$ was expressed as in Equation (2):

$$\frac{V(s)}{F(s)} = \frac{\omega_n^2}{(s+\omega_n)^2 s} \quad (2)$$

where *V(s)*, the output voltage, *F(s)*, applied force to the bimorph.

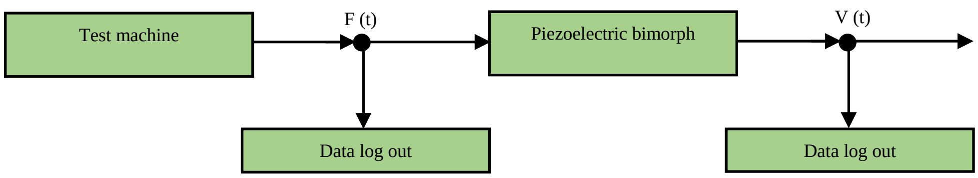
The inverse Laplace transform of Equation (1) may be written in the time domain as in Equation (3):

$$c(t)=1 - e^{-\omega_n t}(1+\omega_n t), \text{ for } t \geq 0 \quad (3)$$

Equation (3) is essential to characterize the dynamics of the piezoelectric bimorph, that is useful at the overall closed loop system control.

The dynamic input wave forms were generated by the 5800 series Instron machine, that includes Advanced Cyclic WaveMaker Software®, that could generate sine and square waveforms [32]. The response was acquired by a DAQ system (NI USB 9221, National Instruments®, Austin, TX, USA) for further processing. The dynamic characteristics namely the frequency response, response time, and damping have been highlighted to evaluate the bimorph behavior [33]. In order to estimate the operating frequencies of the piezoelectric bimorph, a sinusoidal wave was chosen as an input to validate the transient characteristics of the bimorph. Then, the frequency response curves due to the change of the input frequencies were plotted. The diagram to perform the frequency response is shown in Figure 3. The frequency response was tested with different force levels to predict the bandwidth of the bimorph.

Figure 3. Block diagram shows the procedure of measuring the frequency response of the piezoelectric bimorph.



Labview software was utilized to process the acquired data from the NI USB 9221 DAQ system. The overall procedure of acquiring the data was established. To obtain the whole set of data during the calibration procedure, an interface was developed using Labview software to save the data for post-processing.

2.3. Theoretical Calculation of Loads at the Knee Joint

To measure the interface pressure at the lower limb prosthesis, the mechanical concept of forces and moments were calculated. Basically, forces and moments that are present at a prosthetic device are generated due to the contact with the ground. These forces and moments transferred to interface the amputee. The dynamic analysis is basically based on Newton's second law, with the calculation of the forces and moments [34]. Figure 4 shows the diagram of forces and moments relative to the x, y, z axes.

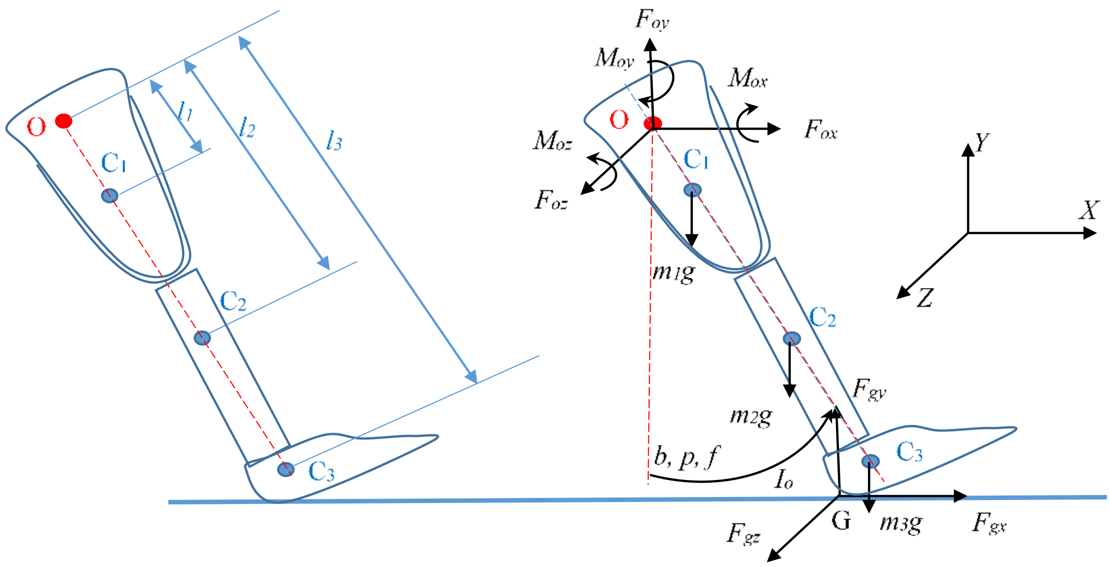
Equation (4) shows the knee rotation by the sum of moments with respect to the origin *O* (Figure 4):

$$M_{oz} - m_1 g_1 l_1 sin(b) - m_2 g_2 l_2 sin(b) - m_3 g_3 l_3 sin(b) + F_{gx} y_g + F_{gy} x_g = I_o f \quad (4)$$

where, b is the angular displacement on sagittal plane; m_i (i = 1, 2, 3) are the stump masses, socket with the tube and prosthetic foot respectively; l_i (i = 1, 2, 3) are the distances from the center of mass to the origin O. Equation (5) shows the sum of moments x regarding O:

$$M_{ox}+F_{gz}y_g+F_{gy}z_g=0 \tag{5}$$

Figure 4. Free body diagram of forces and moments during prosthetic leg heel strike [34].



Equation (6) shows the sum of moments y regarding O:

$$M_{oy}+F_{gz}x_g+F_{gx}z_g=0 \tag{6}$$

Equations (7)–(9) show the sum of forces on the different coordinated axis (x, y, z):

$$F_{ox}+F_{gx}=(m_1+m_2+m_3)\left(rfcos(b)\text{-}rp^2sin(b)\right) \tag{7}$$

$$F_{oy}+F_{gy}\text{-}(m_1+m_2+m_3)g=(m_1+m_2+m_3)\left(rfsin(b)\text{-}rp^2cos(b)\right) \tag{8}$$

$$F_{oz}+F_{gz}=0 \tag{9}$$

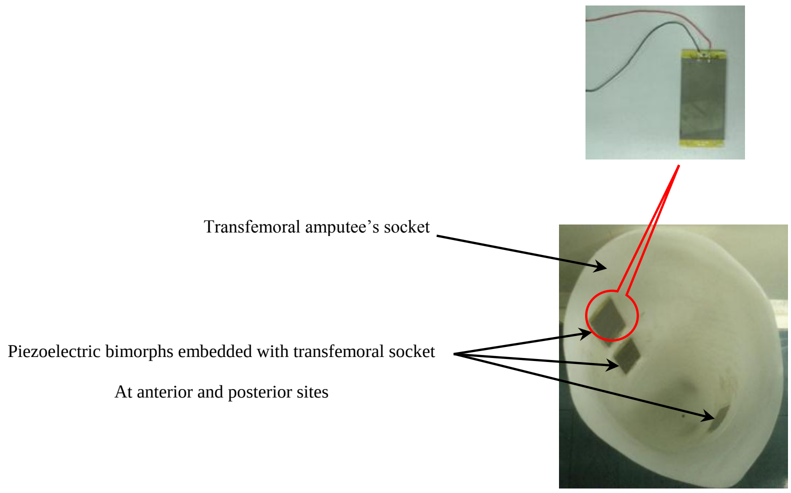
where p is the angular velocity, f is the angular acceleration and r is the distance from the origin to the center of mass of the entire model. The interface pressure is described according to the previous equations. According to Equations (7)–(9), the maximum exerted force was 26.1 N. Based on the calculated values of the maximum exerted force, a sensor that can be used to measure the generated pressure was selected. Therefore, an experimental case of using the piezoelectric bimorph to measure the stump/socket pressure for transfemoral amputees were undertaken.

Transfemoral Subject Trials

To investigate the approach of using a piezoelectric bimorph to detect the stump/socket pressure for a transfemoral amputation subject, an experiment with a single amputee was conducted. The piezoelectric bimorphs were placed in particular socket regions to acquire the maximum amount of pressure from the lower limb that would provide an indication about the gait characteristics. Three piezoelectric bimorphs were embedded to transfemoral amputee's socket as shown in Figure 5.

The session was started by asking a transfemoral amputee subject (age 29 years old, male, 75 kg, height 182 cm) who had been wearing an above knee prosthetic leg for the past 10 years, to perform walking movements at self-selected speed. Piezoelectric bimorphs were inserted in three different locations inside the socket at the anterior proximal, anterior distal, and posterior positions in which the maximum stresses are generated from those sites [22,31,35]. The bimorphs were tethered and the output signals were transmitted via wires to the workstation.

Figure 5. Piezoelectric bimorph sensors embedded to the transfemoral amputee's socket.

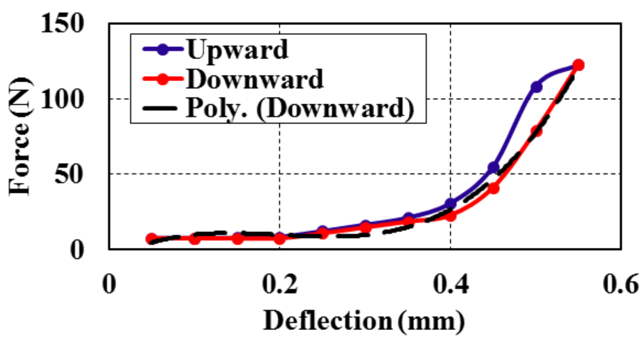


3. Results and Discussion

3.1. Results and Discussion of the Static Test

The static characteristics such as range, linearity, hysteresis, and repeatability were presented to show the bimorph behavior and the operating range at static measurements. Figure 6 shows the output force *versus* the vertical deflection at *z*-direction in reference to Figure 2a. A hysteresis effect was also determined by measuring forces in the upward and downward directions (Figure 6). Figure 7 shows the relation between the input force *versus* the output voltage and the deflection at different values of forces that ranged from 0 to 120 N.

Figure 6. Relation between applied force *versus* the deflection of the piezoelectric bimorph.



The bimorphs' output voltage and deflection can be calculated against the input force value within the range of measurements (Figure 7). The full scale output (FSO) hysteresis of the piezoelectric bimorph at the applied forces during upscale and downscale was calculated and is shown in Figure 8. The sensitivity and linearization can be figured out by plotting the regression line of the piezoelectric bimorph data as shown in Figure 9. The static validation of the piezoelectric bimorph shows a capacity of measuring forces up to 100 N under static operation conditions (Figure 8).

Figure 7. Force *versus* output voltage and deflection.

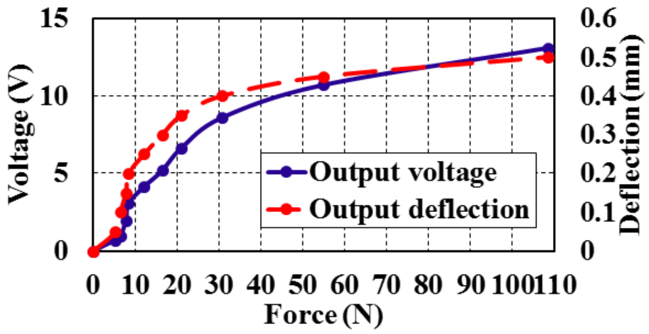


Figure 8. Output voltage from the sensor *versus* the applied force in upward and downward directions.

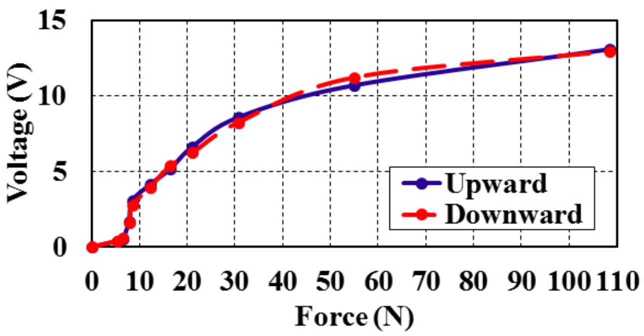
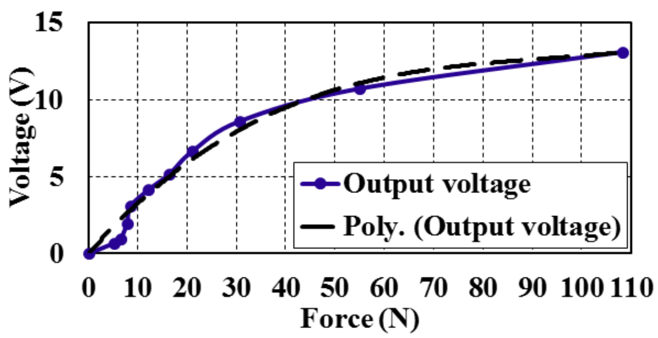


Figure 9. Output voltage of the piezoelectric bimorph *versus* the applied force showing the regression line.



Further comparison between the adopted piezoelectric bimorph and the other existing sensors would be useful to show the differences in terms of the linearity and the range of measurement. Figure 10 shows a comparison that was conducted between the piezoelectric bimorph, FBG sensor, and Flexforce sensors [24,25]. The FBG sensor can measure force and produce s wave length shift that predicts the amount of force. As shown in Figure 10, FBG exhibits an acceptable linearity along the scale of measurements. However, the range of force was smaller compared to the piezoelectric bimorph and Flexforce units. The piezoelectric bimorph has a range of static force measurements of 0–100 N as shown in Figure 10, which is almost same as the Flexforce that has a range of 0–98 N. While both the piezoelectric bimorph and Flexforce have almost a similar range in terms of static force, the piezoelectric bimorph has better dynamic characteristics in terms of dynamic range and operating bandwidth, 0–77 N and 0–35 Hz, respectively (Table 1). However, The Flexforce has a limitation in the dynamic range when it is placed on curved surfaces, as the effective area is bent and this affects the dynamic response of the device [36].

Figure 10. Static force characteristics of piezoelectric bimorph and two different available force sensors.

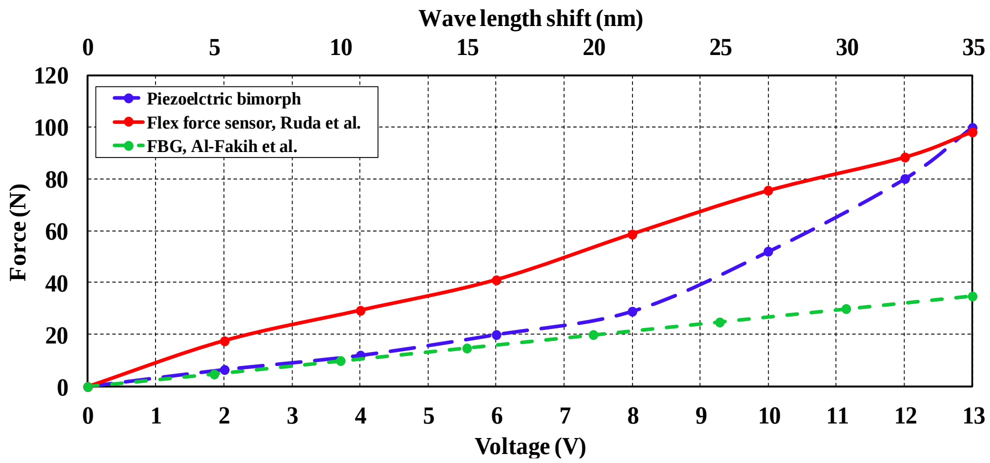


Table 1. Overall characteristics of piezoelectric bimorph.

Characteristic	Value
Average sensitivity	0.3 (V/N) of reading
Linearity error	16.8% FSO
Repeatability	1.1% FSO
Static range	0–100 N
Hysteresis	0.4% FSO
Dynamic range	0–77 N
Operating bandwidth	0–35 Hz
Response time at 95% and 98%	0.22 s at 95% and 0.27 s at 98%
Damping	Overdamped
Overall instrument error and uncertainty	1.9%

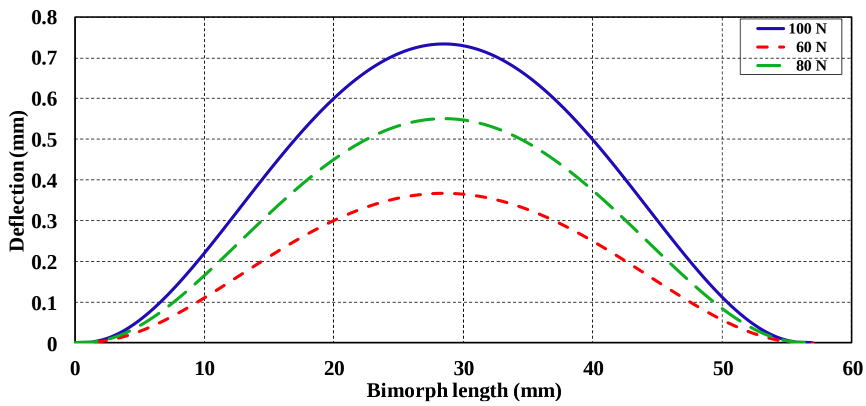
Where FSO, full-scale operating range.

A simulation was conducted using Comsol Software® to illustrate the show the deflection behavior of the bimorph under different applied forces (Figure 11a). The bimorph was deflected according to the amount of load that was applied to the surface area. The forces were applied at 60, 80 and 100 N, respectively, and the bimorph's corresponding deflection along the length was recorded and plotted.

The bimorph showed variations of the deflection values along with different loads, with a maximum deflection of 0.73 mm at 100 N applied force. The maximum values of the deflected bimorph occurred at the middle of the bimorphs length as shown in Figure 11b, thus, the current simulation will assist to better understand the behavior of the bimorph when the real interface pressure of the amputee subject is considered. Furthermore, the surface area of the piezoelectric bimorph (0.001085 mm^2) as shown in Figure 11b provided a wide range of pressure measurement.

Figure 11. Performance of the piezoelectric bimorph, (**a**) Bimorph deflection when load applied at both faces; (**b**) Piezoelctric bimorph's deflection at different applied loads, simulation performed using Comsol software.

(**a**)



(**b**)

3.2. Results and Discussion of the Dynamic Tests

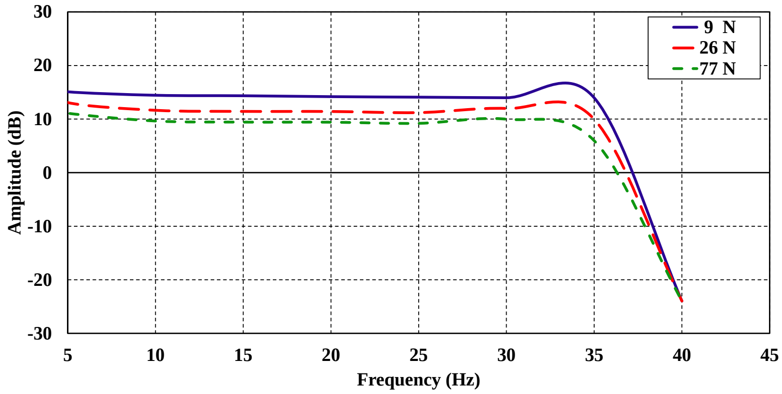
The dynamic characteristics basically show the capability of the piezoelectric bimorph under certain dynamic conditions. In this section, the dynamic behavior of the piezoelectric bimorph is represented. The methods adopted to define the piezoelectric bimorph are namely the frequency response, response time, and damping. These methods were adopted to estimate the dynamic behavior of the piezoelectric bimorph in order to determine the functionality of the device in the field of prosthetic knee development.

3.2.1. Frequency Response

The frequency response is a technique to measure the dynamic response of the piezoelectric bimorph. To obtain the frequency response, a harmonic test function (sinusoidal function) was used as an input signal to the piezoelectric bimorph. The sinusoidal input forces selected were 9 N, 26 N and

77 N, to check the functionality of the bimorph under dynamic conditions. The output was monitored and plotted as shown in Figure 12, presented as the frequency response graph of different applied harmonic forces. In addition, it illustrates the operating frequencies and bandwidth of the piezoelectric bimorph. The frequency investigation shows the capability of the bimorph up to 77 N at dynamic region (Figure 12).

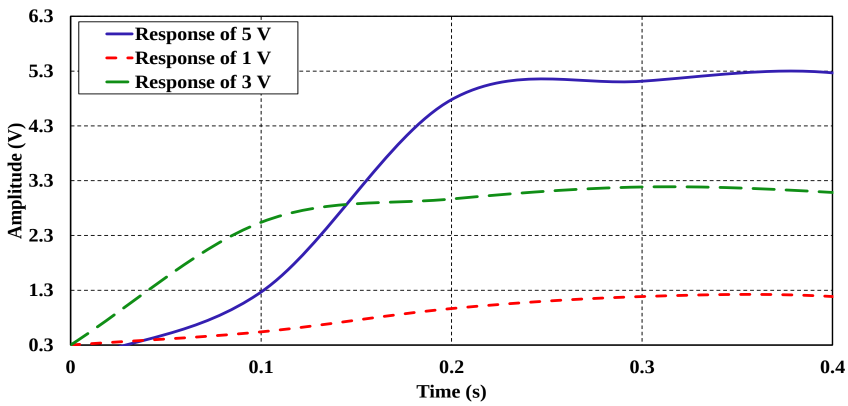
Figure 12. Dynamic response of the piezoelectric bimorph.



3.2.2. Response Time and Damping of the Piezoelectric Bimorph

Response time is another means to define the bimorph's dynamic response. Response time is calculated while the bimorph's output reaches a specific percentage of output value when a step change is applied to its input. The step input function is applied to the system to determine the behavior and speed of the system in response to a change in input. Figure 13 shows the transient response of the voltages that were measured at different levels of 1, 3 and 5 V. Particularly, the 5 V response delivered from the piezoelectric bimorph was selected to calculate its response time at 95% and 98%, respectively. The 5 V response produced response times of about 0.22 s and 0.27 s, respectively, which shows an acceptably rapid response for such a level of voltages.

Figure 13. Sample step responses of the piezoelectric bimorph due to different step inputs.

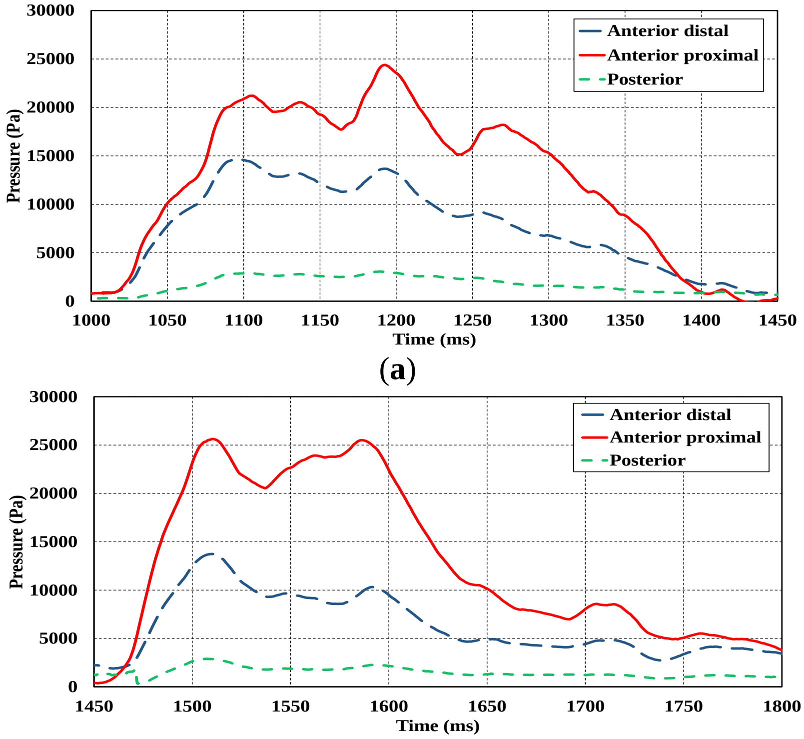


Damping is a sensor's characteristic that defines both how energy from a rapid change in input is dissipated within the bimorph and how it affects the dynamic response characteristics. A critical damping behavior was noticed for the piezoelectric bimorph as can be seen in Figure 13, as it has no overshoot, it is delayed until it reached the final value. An increase in damping in a bimorph may cause the response time and the upper limit of the frequency response to fall. The overall static and dynamic characteristics of the adopted piezoelectric bimorph were obtained and are listed in Table 1.

3.3. Results and Discussion of the Case Study

The results of the pressure distribution inside the socket during the subject trials were presented in Figure 14. Three tests were performed while the subject was wearing a prosthetic knee device. In each gait test, the pressure at the three locations (anterior, proximal, anterior distal, and posterior) was measured. In Figure 14, the pressure distribution was plotted against a time interval of 450 ms each. The average pressure in the anterior proximal region shows a higher amount of pressure during the tests compared to the anterior distal and posterior regions. This agrees with the results of Dumbleton *et al.* [37], although Dumbleton *et al.*, conducted their study on transtibial amputee subjects who wore the socket for daily use for at least 6 months. Zhang *et al.* [32] considered the pressure interface between the stump and the socket by using finite element analysis. Their research revealed that the distribution of the pressure at the anterior region is higher than the posterior region which emphasised the results of the current study. The maximum pressure that was measured at the anterior region was about 25 kPa as can be seen in Figure 14a,b. However, the piezoelectric showed distortion at the measurement level of 30 kPa during the anterior proximal measurement. Due to the internal properties of the piezoelectric material which affect the hysteresis effect, the coupling effect between the mechanical and electrical parameters became saturated during that level of measurements [14,27].

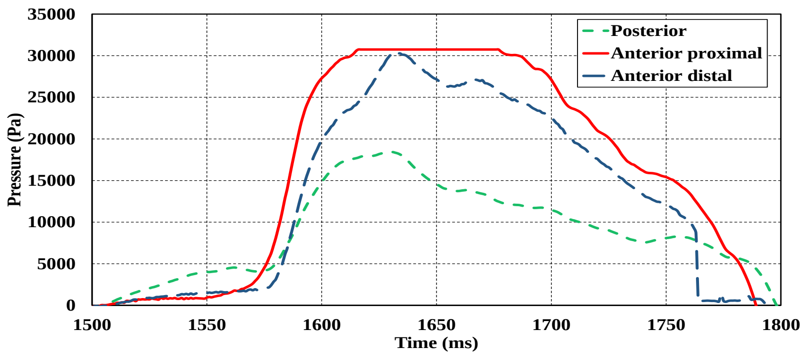
Figure 14. Stump/socket pressure distribution of transfemoral amputee subject during gait, (**a**); (**b**) and (**c**).



(**a**)

(**b**)

Figure 14. *Cont.*



(**c**)

In this work, the overall characteristics of piezoelectric bimorphs were investigated in order to apply them in lower prosthesis development and interface pressure measurement. A case study was considered to present their capability in that field. More specifically, a preliminary measurement of stump/socket pressure of a transfemoral amputee was considered. Gait socket/stump pressure measurements were conducted because of their significant role in prosthesis research.

4. Conclusions

This study was performed to validate the application of piezoelectric bimorph in the prosthetics field. Static and dynamic characteristics of the piezoelectric bimorph were conducted. The dynamic behavior of the bimorph in terms of the response time and bandwidth of operation was investigated. According to the determined characteristics of the piezoelectric bimorph, an assessment of its use as an in-socket sensor was presented. The piezoelectric bimorph sensor was compared to the current Flexforce and FBG sensors in terms of force range and linearity. The piezoelectric bimorph showed similarity to the fFexforce sensor in terms of the static operating range, however the bimorph presented a more suitable dynamic measuring range compared to both the Flexforce and FBG sensors. Furthermore, the current study discussed the usage of the bimorph to measure the interface pressure inside the socket for transfemoral amputee subjects at three different sites. The experiment was conducted with a transfemoral amputee to validate the concept of using the bimorph as a sensing element inside the socket. The results showed that the maximum distribution of the pressure occurs at the anterior region compared to the posterior region. On the other hand at a certain amount of pressure (30 kPa) the signal was truncated due to the saturation of the bimorph's material properties. Thus it can be concluded that the bimorph showed acceptable results for pressure measurements up to 27 kPa and has some limitations for measuring pressures higher than that value. It is recommended to conduct more experiments with subjects of different body weights and pathological considerations to come up with a better understanding of the current approach. Specifically, the measurement of the interface pressure is quite complex due to the combination of normal and shear stress which requires further investigation.

Overall, the preliminary results gathered from the experiments reported in this paper were promising at this stage of research and provided indication about the consistency of the piezoelectric bimorph signals under real measurement conditions. However, more clinical trials utilizing the approach presented in this paper should be performed to validate the capability of the bimorph to measure the shear stresses for both transtibial and transfemoral amputees. Also, more clinical trials with subjects of different weight and level of amputation are recommended. In addition, different activity movements such as sit to stand, slope climbing, and stair ascent/descent could provide further validation of the current concept. Finally, collecting data during clinical experiments could be easier by using a wireless system that facilitates the movement of the subject and provides better handling of the collected data.

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Author Contributions

Amr M. El-Sayed studied the concept, conducted the experiments, analyzed the data, and drafted the manuscript. Nur Azah revised the manuscript. Noor Azuan was responsible for study supervision.

Conflicts of Interest

The authors declare no conflict of interest.

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Article

Development of a Micro-Gripper Using Piezoelectric Bimorphs

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Abstract: Piezoelectric bimorphs have been used as a micro-gripper in many applications, but the system might be complex and the response performance might not have been fully characterized. In this study the dynamic characteristics of bending piezoelectric bimorphs actuators were theoretically and experimentally investigated for micro-gripping applications in terms of deflection along the length, transient response, and frequency response with varying driving voltages and driving signals. In addition, the implementation of a parallel micro-gripper using bending piezoelectric bimorphs was presented. Both fingers were actuated separately to perform mini object handling. The bending piezoelectric bimorphs were fixed as cantilevers and individually driven using a high voltage amplifier and the bimorph deflection was measured using a non contact proximity sensor attached at the tip of one finger. The micro-gripper could perform precise micro-manipulation tasks and could handle objects down to 50 µm in size. This eliminates the need for external actuator extension of the microgripper as the grasping action was achieved directly with the piezoelectric bimorph, thus minimizing the weight and the complexity of the micro-gripper.

Keywords: piezoelectric actuator; bimorph

1. Introduction

Piezoelectric materials are ideal for systems such as micro-grippers due to their fast reaction time and miniaturization potential [1]. They have been used as sensor and actuator components due to their unique reversible electrical and mechanical properties. Applications of piezoelectric materials range from buzzers to diesel engines, fuel injectors, sonar, ultrasound, and nanopositioners in scanning microscopes [2]. As actuators, piezoelectric materials are increasingly important in the latest positioning technology due to their precise displacement [3] and their several other advantages such as quick response, large generative force, and high electromechanical coupling [4]. Piezoelectric actuators are categorized into two configurations: stack actuators and bending actuators. By stacking the piezoelectric layers on top of one another, the cumulative volume of piezoceramics increases the energy delivered to a load. On the other hand, bending actuators consist of multilayers of piezoceramics with greater length than the stacked type. Those multilayers can either be double mounted or single ended as a cantilever [5]. A special case of multilayer bending actuators is the piezoelectric bimorph actuator, which consists of two layers of piezoelectric material connected over their length surfaces. When electric voltage is applied, one layer extends and the other contracts [6,7]. The resultant bending motion becomes the working principle in micro-mechanical applications [8]. Consequently piezoelectric bimorphs have been involved in areas related to precision position control, loudspeakers, vibration damping, noise control, relays, phonograph pick-up, acoustics, and pressure sensing [9].

An important characteristic of bending piezoelectric bimorphs is that the deflection of the bender's tip is dependent on an alternating driving voltage. Many studies have investigated the behavior of piezoelectric actuators. Other studies performed investigations on the nonlinear behavior of bending piezoelectric bimorphs structures under exposure to high electric fields [10], modeling of asymmetrical bending piezoelectric bimorphs structures and the static behavior of the expected bending moment [11], and analytical description of the bending piezoelectric bimorphs' free tip deflection by matrix calculus [12]. The universal deformation state equations were further extended to trimorph bending structures [13]. The free tip deflection of piezoelectric multilayer beam bending actuators under the influence of an electric load was presented by DeVoe and Pisano [14]. The dynamic behavior of a bimorph bending structure excited to bending vibrations by external harmonic forces, bending moments, pressure loads and electrical driving voltages including a flexible plate attached at the free bender's tip had also been established [15,16] and a system of differential equations describing the dynamics of a bimorph was formulated [17]. These establishments of piezoelectric responses contributed towards its application as a micro-manipulating system.

A single-degree-of-freedom micro-manipulator suitable for space robots applications requires lightweight, simplicity, and immunity from magnetic fields [18]. Space robots most commonly require components that can survive at least the rigors of the space and perform exploration, construction, or other tasks. Smart materials are needed for developing some essential parts in space robots for specific applications. For example, robotic hands are used to contact worksite elements safely, quickly, and accurately without accidentally contacting unintended objects or imparting excessive forces beyond those needed for the task. All these tasks require smart materials with minimal time delay to allow distant humans to effectively command the robot to do useful work [19]. End effectors of space based robots must also be dexterous and precisely manage the position of the grasped object. Therefore, a bending piezoelectric bimorph is an ideal solution for an end-effector that could perform such pick-and-place tasks which is the essence of micro-manipulation [20].

Grasping and moving small objects from one location to another depends on the shape and weight of the object, and whether the object is fragile or firm [21]. The basic operation of the micro-gripper depends on the mechanism of the specific type of actuators employed, such as thermo-piezoelectric actuator [20]. The utilized micro-gripper was developed of two parallel lead zirconate titanate (PZT) layers with a fixed range of displacement. Other micro-manipulation mechanisms were designed to enable the tip of the micro-gripper to move in parallel [22]. Static characteristics and control of the micro-manipulator and variation of deflection with the frequency were also reported [1]. This work aims to extend the investigation on parallel micro-gripping in terms of the effect of the sandwiched supporting layer on deflection of the piezoelectric bimorph actuator and to understand to what extent the rigidity of the bimorph varies due to the brass layer between both piezoceramic layers.

The second aim is in light of the implementation of parallel micro-gripper by utilizing the piezoelectric bimorph itself to grasp soft objects instead of attaching additional flexible cantilever [20,21]. It has been well reported that piezoelectric was used as an actuator for driving micro-manipulation of micro-objects [1,22,23]. A piezo-actuator was also utilized as a driver to provide movement to the flexible amplification mechanism of the micro-gripper. It may deliver a large force, but the size is big and it has a complex structure. This study aims to achieve the grasping action directly from the piezoelectric bimorph in order to minimize the weight and complexity of the micro-gripper. In addition, the essential role of the supporting brass layer in providing the essential behavior of the micro-gripper as well as increasing its life cycle was to be established.

This article presents: (i) the basics of a piezoelectric bimorph and the equations of deflection for both the non-supporting sandwiched layer and the brass supporting layer, (ii) the experimental setup that was used for piezoelectric bimorph characterization and also the overall diagram of the micro-gripper in addition to the tests performed for micro-gripper validation, and (iii) the theoretical and experimental results and discussion of the bimorph and the micro-gripper characteristics.

2. Configuration of Bending Piezoelectric Bimorphs

Bimorphs, which are commonly used as a fundamental element in many operating devices, are made of two piezoelectric sheets bonded together [24]. They were also used to control the vibration of a helicopter rotor blade with limited success [25]. Relations between intensive parameters, which refers to the deflection, bending angle, volume displacement and electrical charge derived for any point over the entire length of the piezoelectric bimorph; and extensive parameters, which refers to variables such as force, bending moment, pressure load and electrical driving voltage [26] have long been established [6,7,10,13,15]. The basic geometry, dimensions, and extensive parameters of our bimorph actuator are shown in Figure 1 where two piezoelectric layers are bonded together with the same polarization. After applying an electric field the piezoelectric bimorph will be deflected as shown in Figure 2.

Figure 1. Dimensions, extensive parameters, and polarization of the bimorph actuator.

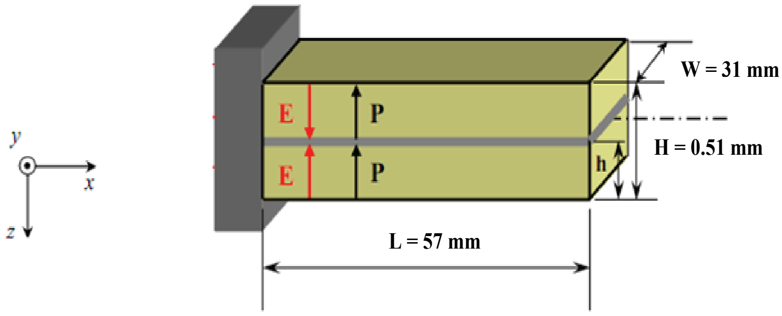
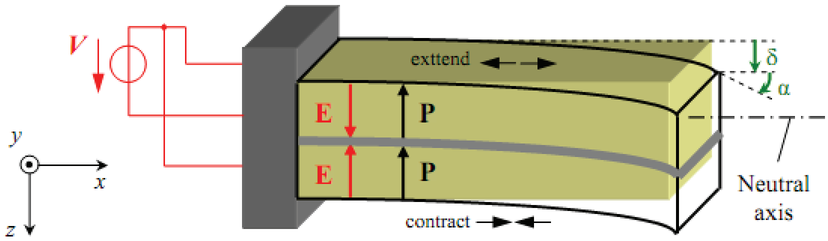


Figure 2. Basic intensive parameters of bimorph actuators after applying electric field.



Generally, piezoelectric bimorph layers are bonded together with a sandwiched supporting material. To establish the relevance of the supporting material, two types of piezoelectric bimorph were employed, in which one of them had the sandwiched supporting material removed. The relationship between the exciting voltage and the output deflection was estimated in both cases [10] and is discussed herein using the equations of deflection *versus* the applied voltage.

Case 1: Without supporting material between the two piezoelectric layers.

The two layers are identically in geometrical, electrical, and thermal parameters. The analytical bending curvature is given by Equation (1) [10]:

$$\delta(x) = \frac{3d_{31}x^2}{H^2}V \tag{1}$$

where $\delta(x)$, the deflection at any position -x (mm)

d_{31}, piezoelectric coefficient (mm/V)

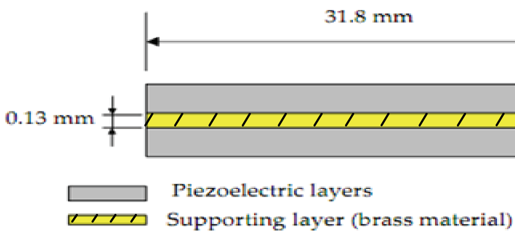
H, total hickness of piezoelectric actuator (mm)

V, applied voltage (V)

Case 2: With a supporting brass layer between the two piezoelectric layers.

The typical configuration in this case of bimorph, which consists of a thin brass metal substrate sandwiched between two piezoceramic patches is presented in Figure 3.

Figure 3. Cross sectional area of the used bimorph actuator.



Equation (2) [10] shows the constitutive relationship of the triple layer piezoelectric bender with applied electric voltage V and tip deflection δ:

$$\delta = \frac{6 s_{11}^{m} d_{31} \left(h_m + h_p \right) L^2}{2 s_{11}^{m} \left(3 h_m^2 h_p + 6 h_m h_p^2 + 4 h_p^3 \right) + s_{11}^{E} h_m^3} V \quad (2)$$

where $S^m{}_{11}$, elastic coefficient of the supporting layer (m^2/N)

$S^E{}_{11}$, elastic coefficient of the piezoelectric layer (m^2/N)

h_m, thickness of the supporting layer (mm)

h_p, thickness of the piezoelectric layer (mm)

L, length of the piezoelectric bimorph (mm)

3. Experimental Setup

Experimental setup consists of two parts. The first setup was for measurement of static and dynamic characteristics of the piezoelectric bimorph (Figure 4) while the second setup is the general layout of the developed micro-gripper based on the previously obtained characteristics (Figure 5).

Figure 4. The experimental setup showing the sensor and piezoelectric actuator amongst other components.

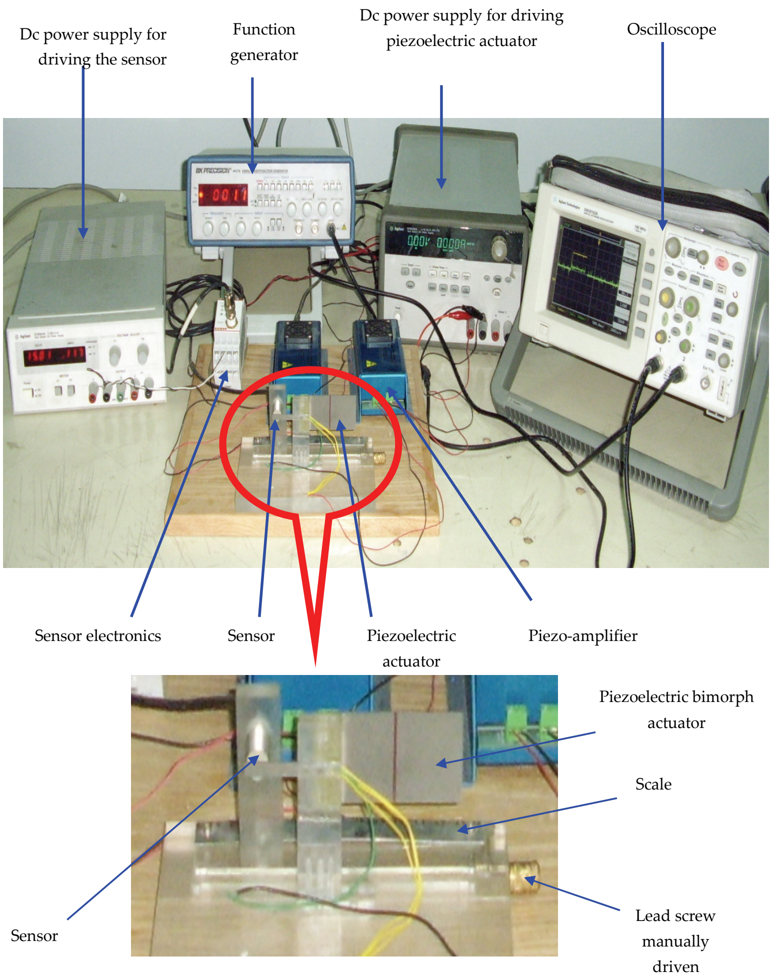
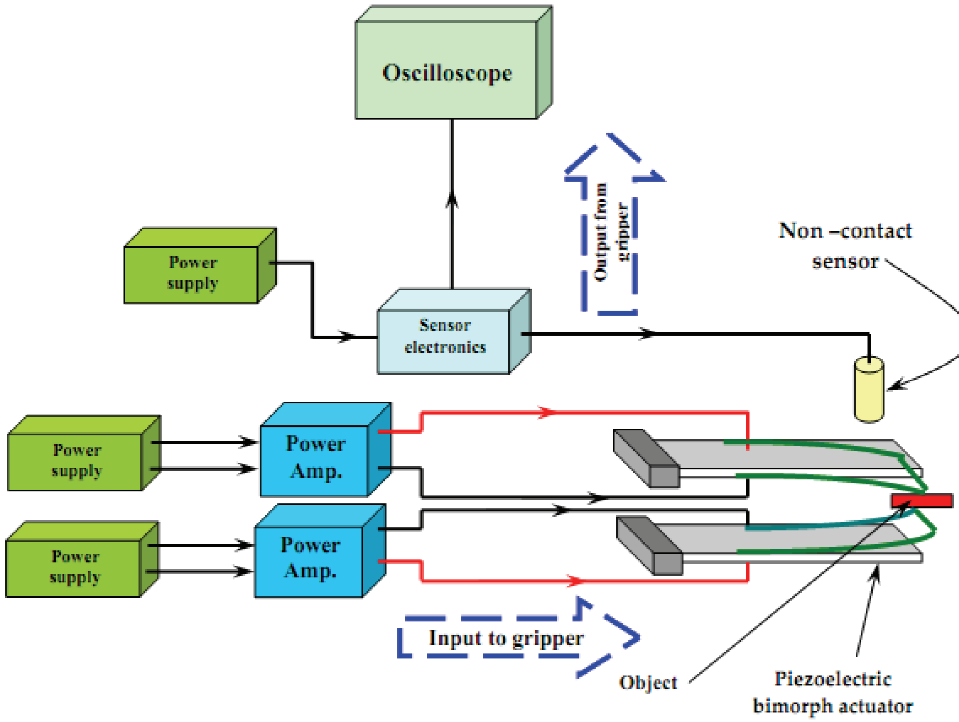


Figure 5. General layout of the developed gripper.



3.1. Part 1: Static and Dynamic Characteristics Measurement

The apparatus consisted of a bending piezoelectric bimorph of two layers of piezoelectric material bonded together with opposite polarity in the form of a cantilever beam (Figure 4). The bimorph was connected to a mechanical breadboard and driven by a piezo-linear amplifier (Model EPA 007). The EPA-007 was a compact high voltage linear non-inverting amplifier, which was used as a high voltage driving source for the piezoelectric actuating device. The bimorph position was measured using a commercial high-resolution capacitive position sensor mounted on a carriage moved with a lead screw. A DC power supply and function generator were used to generate the drive voltages for the piezo driver. The application of an electric field to the bimorph caused one layer to extend slightly and the other layer to contract slightly in the x-direction. The differential length caused the beam to bend towards the contracting layer. The movement of the cantilever was adjusted by precisely regulating the applied electric field.

3.2. Part 2: General Layout of the Developed Micro-Gripper

The developed micro-gripper (Figure 5) consisted mainly of two piezoelectric bending bimorphs, *i.e.*, fingers. Both consisted of two PZT layers in which the bimorph was actuated by applying an electrical voltage across its width. A linear amplifier with a supply input signal drove each finger individually. By applying specific voltage to the bimorph, a definite proportional deflection was produced. The deflection was measured using a non-contact proximity displacement sensor. The output signal from the sensor was displayed on a digital oscilloscope.

Figure 6 illustrates a two-fingers parallel micro-gripper with a position sensor attached at the tip of the finger. The maximum displacement of the actuator was approximately 2,000 µm. The resulting gripper displacement was sufficient to grasp mini objects. To compensate for the small displacement of the finger, one of the fingers was fixed and the other was mounted on a carriage moved by a lead screw of 1,000 µm resolution. Two different objects were used to test the performance of the micro-gripper. Object 1 was a thin strain gauge (17.5 mm × 7.5 mm × 0.1 mm) and object 2 was a smaller strain gauge (8 mm × 3 mm × 50 µm). Figure 7 shows both objects used for assessment of the micro-gripper performance.

Figure 6. Front view of the two parallel piezoelectric bimorph with the non- contact sensor.

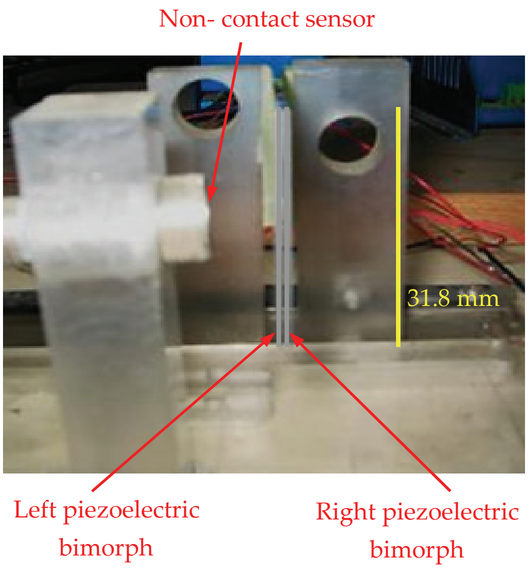
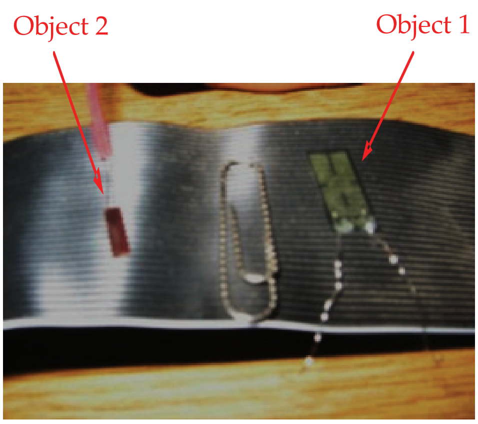


Figure 7. Two types of strain gauges (selected as a micro-objects).



Experiments were performed by grasping object 1, *i.e.*, the strain gauge, as shown in Figure 8. A gap of 100 µm was produced by connecting both fingers with the same voltage. Then, another validation test was performed by picking up object 2 as shown in Figure 9. The micro-gripper successfully grasped the two different objects of different sizes. In the same manner, other objects with various sizes could be manipulated by setting the range of the gap between the two parallel bimorph fingers.

Figure 8. The micro-gripper grasps small object 1 (strain gauge).

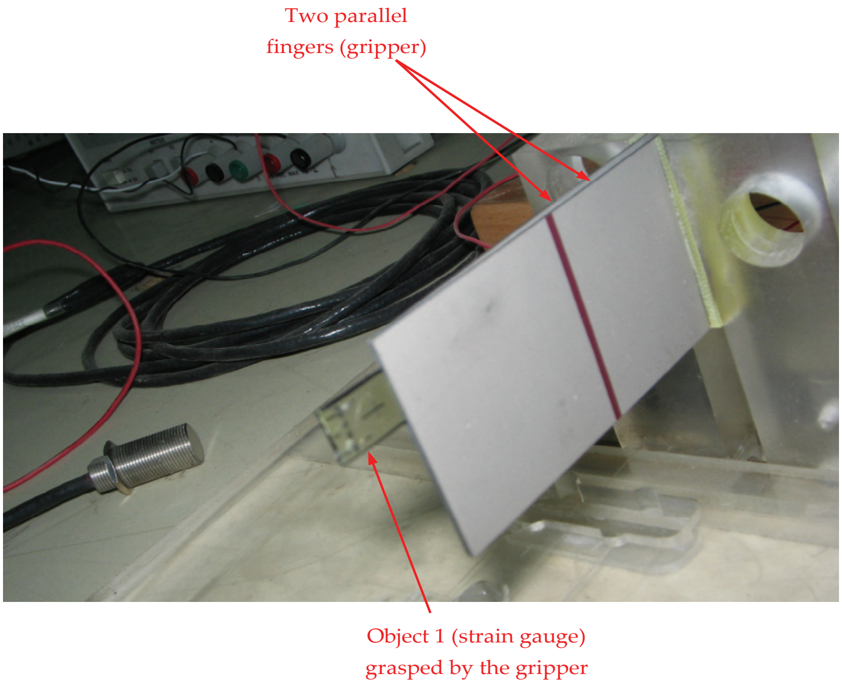
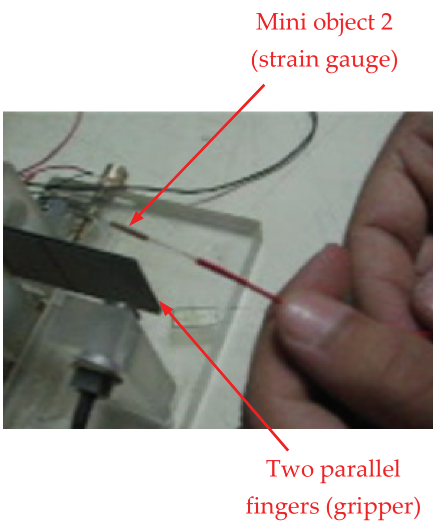


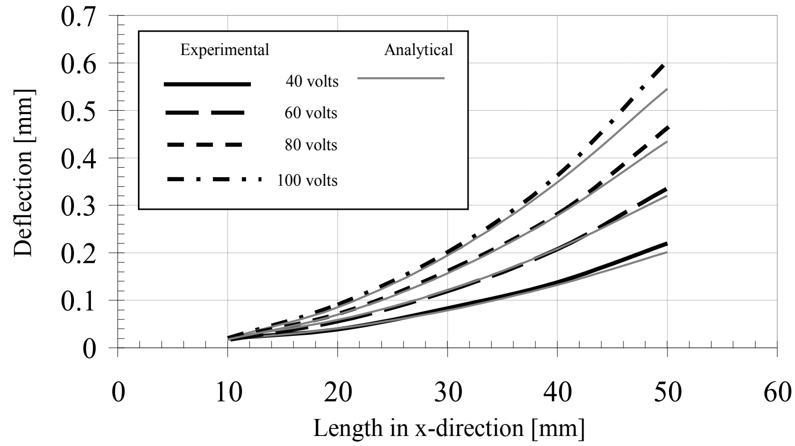
Figure 9. The developed micro-gripper carrying strain gauge (micro-object).



4. Results and Discussion

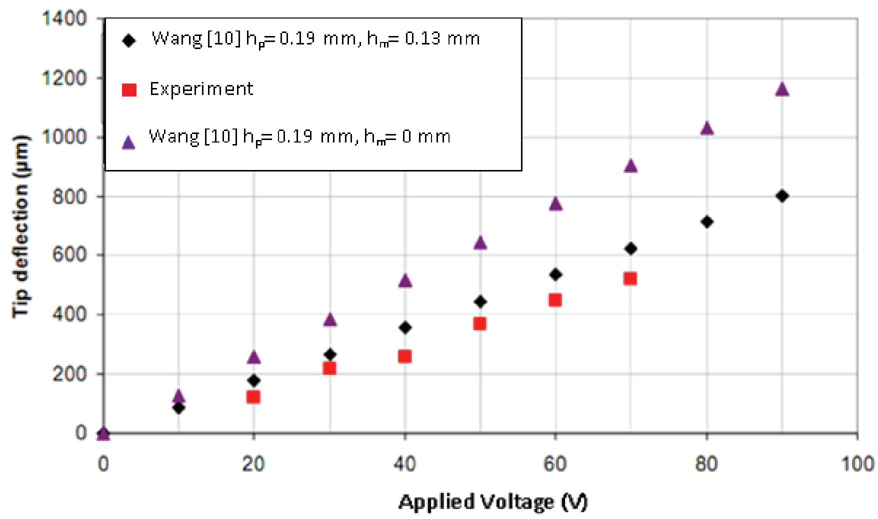
The characteristics of the developed micro-gripper in terms of all possible responses of the piezoelectric bimorphs were presented by measuring the bimorph's deflection. Variations of the measured deflections of the bimorph actuator along the length of the actuator for different driving voltages are illustrated in Figure 10. The experimental investigation was employed by varying the excitation voltage of the bimorph actuator. A bimorph actuator of 57.2 mm length was used for estimating the overall characteristics. The experimental results are correlated with the analytical results within an acceptable error of approximately 2%.

Figure 10. Experimental and analytical deflection of the bimorph according to different applied voltages.



Further assessment of the tip deflection is presented in Figure 11, which highlighted the performance of experimental result conforming to theoretical values in case 2 where the bimorph is sandwiched with a brass layer of a thickness 0.13 mm.

Figure 11. Tip deflection *versus* applied voltage of piezoelectric bimorph, h_p: thickness of the piezoelectric layer (mm), h_m: thickness of the supporting layer (mm).



Nevertheless the results of case 1, *i.e.*, without supporting brass layer, did not agree with the analytical results. This verified the statement regarding the rigidity provided by the supporting layer. From the theoretical calculations of Equation (2) it was confirmed that the bending piezoelectric bimorph without the supporting layer produced a greater tip deflection of up to 1,150 µm, as illustrated in Figure 11.

4.1. System Resolution and Sensitivity

The observation of either the force or displacement at the tip of the piezoelectric bimorph might be assessed based on the characteristics of the grasped object. The study assumed that the object to be gripped was delicate and lightweight, thus no slipping occurred during the handling process. Therefore, the assessment was performed for the deflection of piezoelectric bimorph and both resolution and sensitivity were considered to characterize the utilized bimorph.

Resolution of this system was defined as the output displacement of the device corresponding to the input voltage [27–29]. The resolution of the piezoelectric bimorph was dependent on the sensor used to measure the resulting displacement [30]. Results showed the resolution of the utilized piezoelectric bimorph finger for micro-positioning actions to be about 80 µm, based on the non-contact proximity sensor used in the microgripper system. The resolution in terms of grasping action ability, defined by the thickness of the smallest object the gripper can grasp, was demonstrated through the experiment of grasping object 2 in this case, *i.e.*, 50 µm, which was based on the smallest gap between the fingers.

Sensitivity indicates the amount of change in the output, *i.e.*, the displacement of the bimorph, as a result of change in the input, *i.e.*, the excited voltage [27–29]. The sensitivity in the current application was determined from the gradient of the output displacement *versus* input voltage graph (Figure 11) through experimental investigation to be 4 µm/V. The relationship between the input voltage, V, and the blocking force, F, was 2.5 mN/V, derived from a theoretical relationship [10] as shown in Figure 12. The frequency curve of bending piezoelectric bimorph *versus* the length is shown in Figure 13, in which the rate of the natural frequency using bimorph of 57 mm in length is about 70 Hz.

Figure 12. Blocking force at the tip of piezoelectric bimorph *versus* applied voltage.

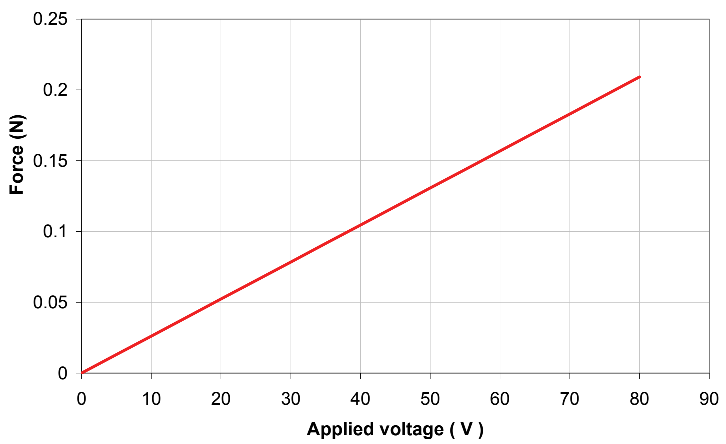
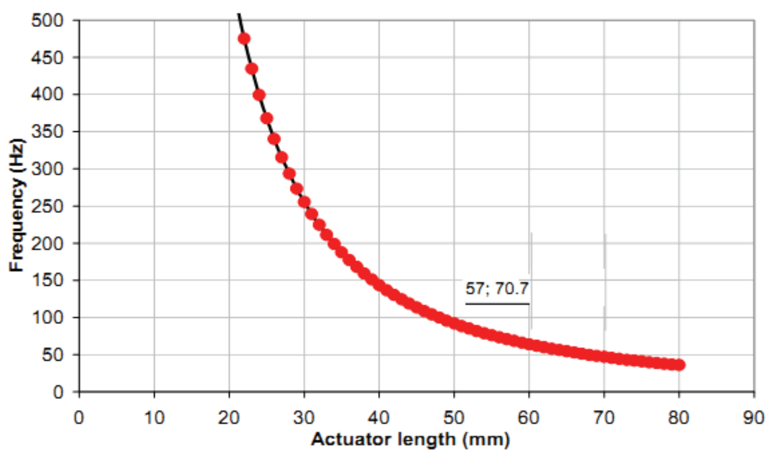


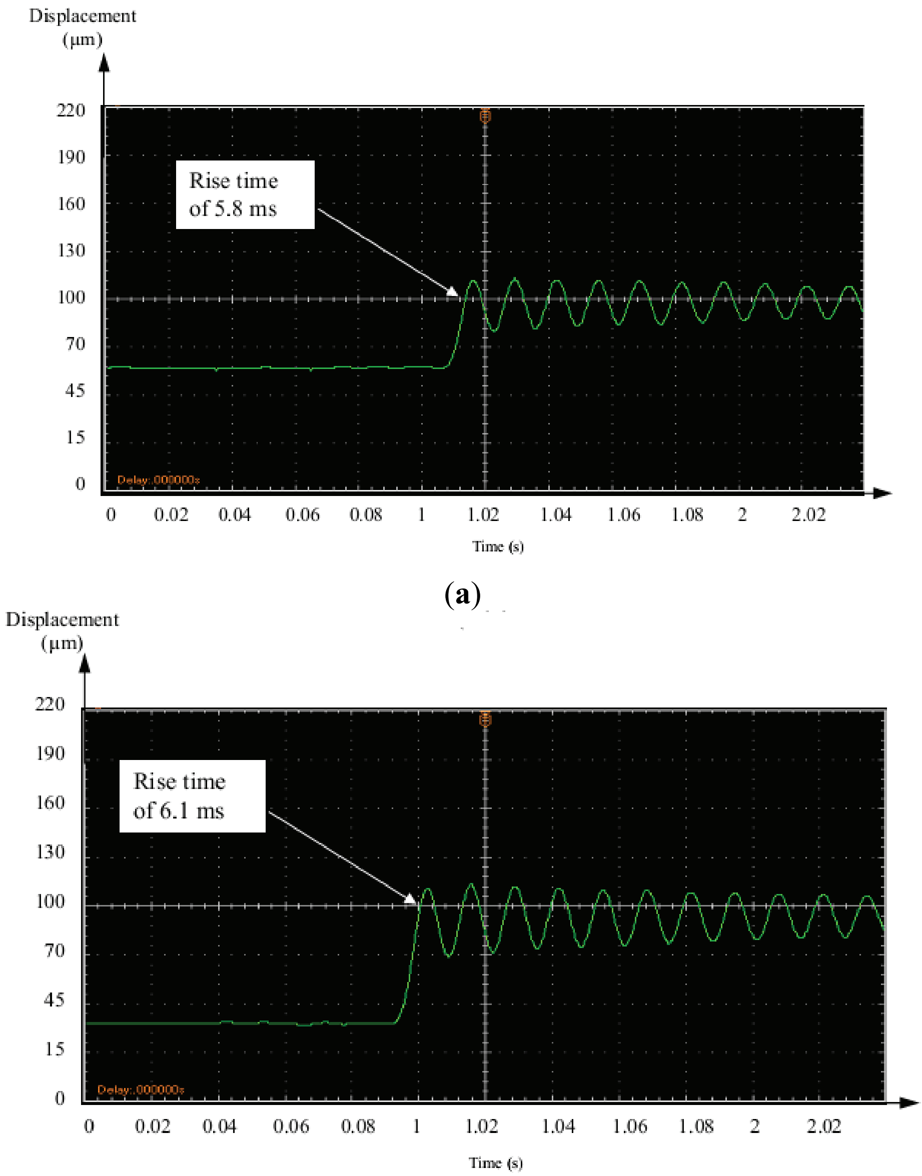
Figure 13. Relationship between frequency and actuator length.



4.2. Step Responses

The step displacement responses of the piezoelectric bimorph due to different input amplitudes of 40 µm and 70 µm are illustrated in Figure 14, respectively.

Figure 14. Transient response due to different step voltages. (**a**) 40 µm; (**b**) 70 µm.



(**a**)

(**b**)

The results indicated a fast response time of 0.05 s. However, the rise time was about 5.8 ms and 6.1 ms respectively. This satisfied the micro-gripper specification of control performance.

5. Dynamic Response of the Piezoelectric Bimorph

Figures 15 and 16 illustrate the dynamic displacement response under different AC driving signals of the actuator (1 Hz, 20 V).

Figure 15. Sine-wave signal.

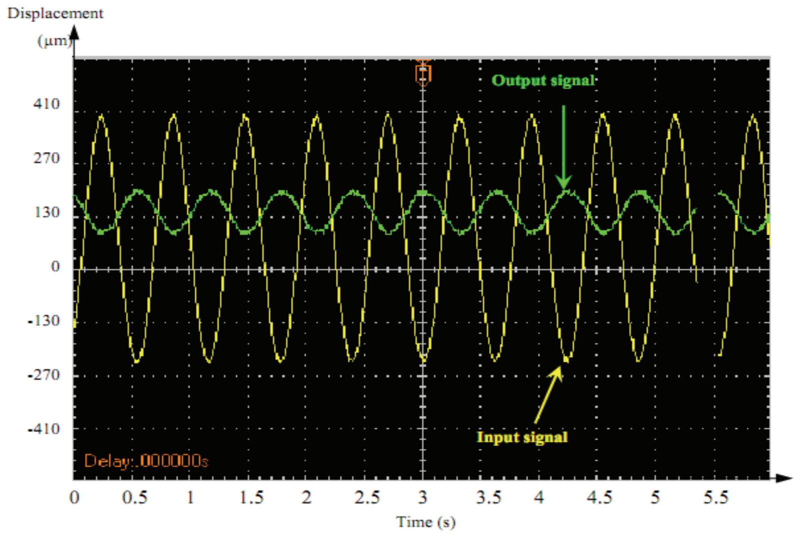
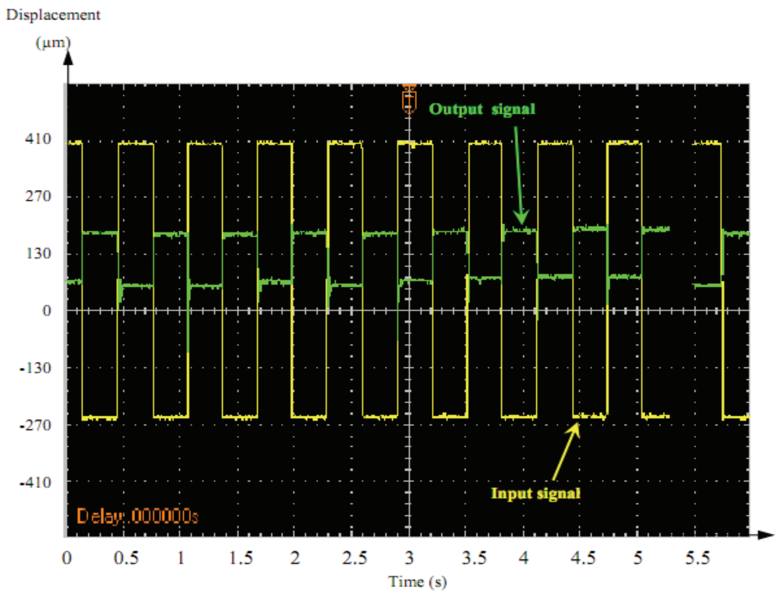


Figure 16. Square-wave signal.

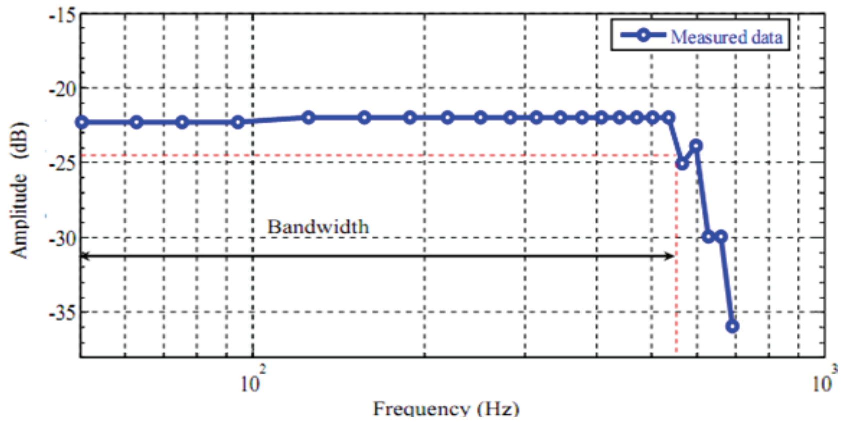


The output response of the square-wave was characterized by vibration followed by overshoot at the front edge of the square–wave driving voltage. A sine-wave output signal showed an acceptable response compared to square-wave signals. Therefore, sine–wave was better adopted as the control signal for dynamic precision positioning.

6. Frequency Response Measurement

The frequency response of the tip displacement of the piezoelectric bimorph was measured with an input signal of 5 V in the frequency range: 0.2–110 Hz and the results are shown in Figure 17.

Figure 17. Variation of the deflection with the frequency for bimorph actuator gripper.



The bandwidth estimated from the obtained results was 101 Hz for the bimorph actuator. The peaks and valleys show that the piezoelectric bimorph can be described as an underdamped system.

7. Conclusions

The characteristics of piezoelectric bimorphs bending actuators were obtained and the experimental setup of the piezoelectric actuator bimorph was successfully developed. The micro-gripper was developed using two parallel piezoelectric bimorphs (fingers) with a non-contact position sensor at the tip of one finger. Each finger was essentially a bending piezoelectric bimorph. To compensate the small displacement of the finger one of the fingers was fixed and the other was supported on a carriage moving on a lead screw, therefore, a sufficient range of object sizes can be handled by changing the initial distance between the fingers. Two micro objects of different dimensions were used as objects to check the validity of the developed micro-gripper. The piezoelectric bimorph micro-gripper time response was reasonable for precise engineering applications with a sine–wave being recommended as the control signal for dynamic precision positioning. This study provided a holistic characterization of a microgripper system for closed loop control as well as the utilization of the piezoelectric material itself as the gripper without requiring additional extension. To measure the micro-gripper displacement and blocking force more advanced sensors with higher accuracy are needed. Further investigations could be undertaken to assess the micro-gripper displacement control and to further develop the micro-gripper to perform more than a single axis movement for advanced automated applications.

Conflicts of Interests

The authors declare no conflict of interest.

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Research Article

Detection of Prosthetic Knee Movement Phases via In-Socket Sensors: A Feasibility Study

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This paper presents an approach of identifying prosthetic knee movements through pattern recognition of mechanical responses at the internal socket's wall. A quadrilateral double socket was custom made and instrumented with two force sensing resistors (FSR) attached to specific anterior and posterior sites of the socket's wall. A second setup was established by attaching three piezoelectric sensors at the anterior distal, anterior proximal, and posterior sites. Gait cycle and locomotion movements such as stair ascent and sit to stand were adopted to characterize the validity of the technique. FSR and piezoelectric outputs were measured with reference to the knee angle during each phase. Piezoelectric sensors could identify the movement of midswing and terminal swing, pre-full standing, pull-up at gait, sit to stand, and stair ascent. In contrast, FSR could estimate the gait cycle stance and swing phases and identify the pre-full standing at sit to stand. FSR showed less variation during sit to stand and stair ascent to sensitively represent the different movement states. The study highlighted the capacity of using in-socket sensors for knee movement identification. In addition, it validated the efficacy of the system and warrants further investigation with more amputee subjects and different sockets types.

1. Introduction

An amputee user's locomotion phase detection in the field of transfemoral prosthesis system is still undergoing extensive research, especially detection originating directly from the users themselves. In general, a transfemoral prosthesis system has always been mechanically based in which the user had to adapt his gait pattern to accommodate the passive behavior of the prosthesis. Having knee joint across the prosthesis increased the complexity of the system but over the years, advancement of passive adaptive and active prosthetic knee has resulted in improved systems and designs for transfemoral amputees [1–3].

Nowadays, active prosthetic knee systems utilized sensors at certain locations around the prosthetic knee to measure specific parameters. Most of the current sensory systems in the development of prosthetic knee devices are usually located away from the knee axis and the muscles themselves. Such sensors measure parameters such as force, torque, position, velocity, and phase transitions for appropriate control decisions. The information derived from these mechanical sensors was used to derive the instantaneous state of movement to further control the prosthesis system. However, more accurate information about the user's instantaneous state of movement could be derived from the sensors if they are located closest to the user peripherals or nearby the knee joint axis itself. Optimal location of the sensors in a prosthetic knee system may provide better deduction capability of the prosthesis to improve user interaction and performance during daily activities, as the accuracy gained from better sensor placement could reduce the complexity of the knee control.

The identification of the different parameters during prosthetic knee movement is essential to control the knee. For example, the most critical input to be addressed during a transfemoral prosthesis controlled gait is the foot position, either on or off the ground, and this was determined from the angle, torque, and force sensors measurements. As the transition between gait phases is crucial for the control of active knee, such inertial sensors are used to recognize the transitions between the gait phases [4–6]. A magnetorheologic fluid actuated prosthetic knee used a strain gage sensors as an axial force sensors [1], sensor that is placed nearby the knee axis to detect force and torque [1]. The axial force sensors measured the force applied to the prosthetic knee from the ground in the longitudinal direction of the knee. Measurement of the knee torque was conducted by classifying the difference between the signals of the forward and hind strain gages [1]. Other sensory mechanisms used by other developed systems were summarized and presented in Table 1. In general, all sensors were embedded into the prosthetic system to deduce the user's current and intended knee movement without measuring them directly from the socket.

Another approach that is used to gain direct input from the muscle to control the active prosthetic knee is by using the electromyography (EMG) system. Direct user interaction was enabled in an active prosthetic knee system by embedding EMG system. In systems that incorporate EMG, the sensors are positioned to detect the user's flexor and extensor muscles activities from generally the rectus femoris, vastus lateralis, vastus medialis, biceps femoris, and semitendinosus. The EMG signal was utilized to formulate the control algorithm that assists the user to control the torque during activities such as stair ascent. However, the muscle activity may be varied depending on the individual amputee's residual limb muscles or according to the amputation type and level. This may require additional adjustment to the EMG electrodes and the control system [7, 8]. Inertial sensors such as accelerometer and EMG were used in combination to identify the start of the gait by using a technique called "per leading limb condition" of the prosthetic leg during walking [9]. However, skin conditions of the transfemoral amputees may affect the use of EMG [10]. In addition, the placement of EMG onto the skin surface and inside the socket may cause skin irritation and affect the user's comfort [11]. Therefore, another approach is needed to improve the control of the active knee device by choosing proper locations of the sensory system. The suitable location of the sensors could minimize the complexity of the control scheme of the lower prosthesis.

The signals from the inertial sensors are not the only ones that may be acquired to help improve the control of the active knee prosthesis. Further investigation on other alternative signals for characterizing the prosthetic knee movement for better control of the knee prosthesis should be conducted and integrated into future system developments [12]. Alternative options that could better characterize the knee movement will aid the designer to identify multiple solutions to improve the area of active prosthetic knee development [13].

Nowadays, researchers try to involve the amputee subjects with the sensory system more closely to assist the controller decision making. Attempts are ongoing to assist the amputee subjects to interact more naturally with the sensory system by making use of the specific high pressure locations inside the socket. Other sensors placed inside the socket such as the F-socket sensor have been used in investigating the pressure around the residual limb, but they were not meant for daily integration into the socket for identifying knee movements in active transfemoral prosthesis [14]. Various kinds of pressure sensors are used to measure the pressure for both transtibial and transfemoral amputees [14, 15]. Current pressure socket measurement systems such as F-socket (Tekscan, Inc., South Boston, USA) or pressure measuring system (Novel, Germany) were used to cover the circumference of the residual limb. However, they have to include all the posterior, anterior, lateral, and medial compartments of the residual limb. Nevertheless, by selecting specific locations inside the socket, limited number of sensors could be placed to provide sufficient measurements that would help to better improve the control scheme of the active knee.

In general, we proposed that direct user signals could be collected from sensors embedded in the socket and residual limb. This study aims to embed the sensory system inside the patient's socket, as this approach will provide less additional components and practically less setup time, thus more flexibility to the patient wearing the socket. This paper presented the efficacy of embedding mechanical sensors inside the socket's internal wall for movement identification. FSR and piezoelectric sensors were placed inside the socket to achieve the aim of this study. In the proposed study, the obtained in-socket data from the interaction between the sensors and the amputee, as well as the biomechanical position of the ground reaction force acting against the sensors inside the socket due to the amputee's specific body posture, will enable the recognition of the user's leg movement as well as events of the movement. These were done by considering the signals from the sensors at different prosthetic knee movements performed by the amputee subject.

2. Materials and Methods

2.1. Sensor Characteristics and Utilization. The adopted sensors (FSR and piezoelectric) in the current study were placed inside the socket wall (Figure 1). FSR was chosen based on its small size (1.25 mm thickness and 12.7 mm diameter) that will not affect the user comfort. Similarly the piezoelectric sensor has a configuration (Figure 2) as well as dynamic characteristics that make it suitable for such applications [16]. The sensors were tethered to transmit the data directly to the PC via wires. The minimal thickness did not affect the user's natural movements. These sensors were able to accurately characterize the knee movements during walking, stair climbing, and sit to stand.

2.1.1. FSR Sensor and Piezoelectric Sensors. Two FSR sensors (Interlink Electronics 402, Interlink Electronics, USA) of sensing area diameter 12.7 mm were used in the current study based on the site that generated maximum stresses [17]. A signal conditioning circuit was built to acquire the output voltage from the FSR at a range of 0 to 3.5 volts. The output voltage from the FSR circuit was connected to a Simulink environment by using the Real-Time Windows Target Toolbox. Afterwards, a data acquisition system (Advantech PCI-1710HG, Advantech, USA) was utilized to analyze the output data from the FSR sensor.

The FSRs were placed at specific locations in the socket to effectively capture the maximum stress of the socket's

Table 1: Sensory mechanisms used in prosthetic knee systems.

Author (year), system	Sensor type	Mechanism and function
Kapti and Yucenur (2006) [5], artificial knee joint	Rotary knee angle's potentiometer	Detects different angles of the knee joint from 119.5° to 180° as the sensor located at the joint centre.
Sup et al. (2009) [6], Vanderbilt prosthetic leg	Load cell	Detects force and torque loading at the knee and ankle.
	Rotary potentiometer	Detects the knee joint angles.
Martinez et al. (2009), agonist-antagonist prosthetic knee	Rotary encoder Digital encoder, to measure Ankle Angle Digital encoders, to measure motor displacements Hall sensor, to measure springs' Compression Force sensitive resistor, to Heel/Toe Contact	Detects the joint angles by controlling the motor displacement via the rotary encoder, attached to the motor shaft.
Sup et al. (2009) [6], Vanderbilt prosthetic leg	Custom load cell	Custom load cell was made to detect force and torque loading at the knee and ankle.
	Potentiometer	Detects the knee joint angles.
Geng et al. (2010) [4], four-bar linkage prosthetic knee	Knee angle sensor used to detect angle at different phases.	Prosthetic knee with four-bar linkages mechanism

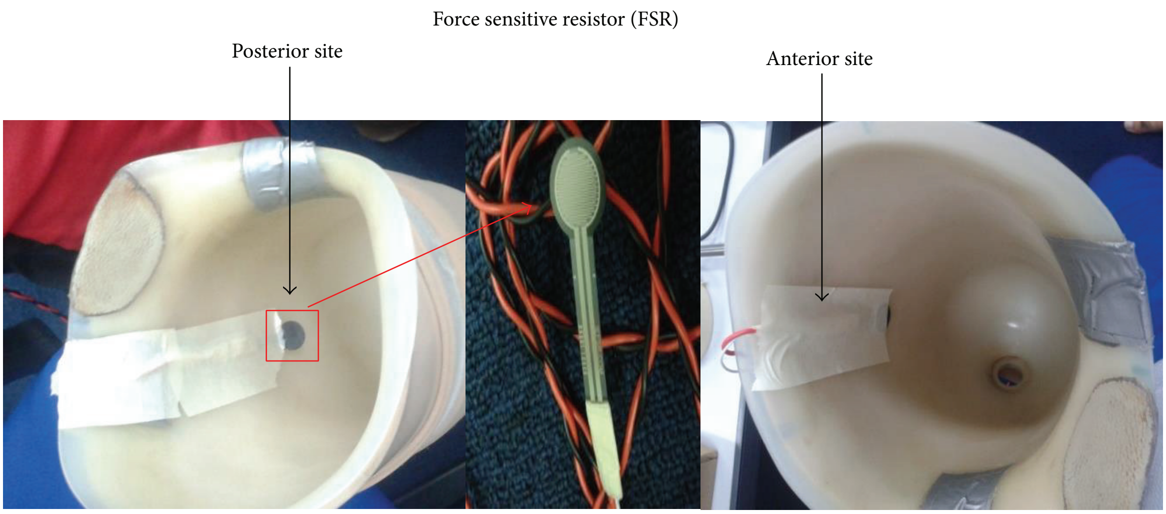

Figure 1: FSRs locations inside the socket during the experiment for both anterior and posterior sites.

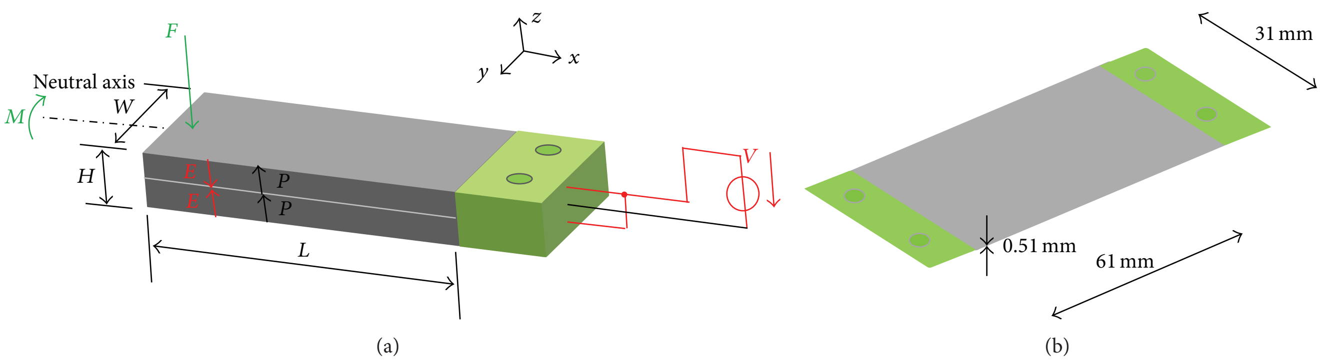

Figure 2: (a) Basic dimensions, extensive parameters, polarization, and applied electric field acting on the bimorph generator [16]. (b) Dimensions of the used bimorph with two fixed ends.

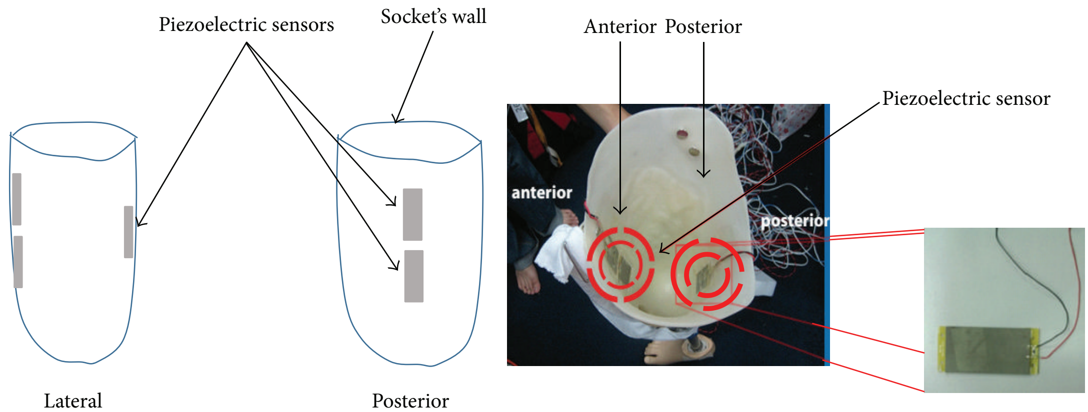

Figure 3: Placement of the piezoelectric sensors at both anterior and posterior sites.

area [15]. Given the small area covered by the sensor, the anatomical muscle bulge during maximum contraction was identified to determine the sensor placement in the socket. Furthermore, to ensure that the sensor is in contact with the greatest pressure point against the socket wall when the muscle contracts, investigators palpated the muscles during maximum voluntary contraction of the amputee's residual limb. This ensured that the FSR was located at a position that allowed detection of highest variation of the signal originating from high pressure at the rectus femoris and biceps femoris muscles contraction [15, 18]. Given the minimal thickness of the FSR (<1.25 mm), the FSR was secured using adhesive sticker inside the socket's wall. This eliminated the user sensational awareness about the FSR in the socket which otherwise would have affected the user's natural movements. Trials were conducted to estimate the pattern variation of three major movements, namely, (i) full stance of gait, comprising heel strike, flat foot, and toe off; (ii) stair ascent; and (iii) sit to stand. The socket with the attached in-socket FSR is presented in Figure 1.

Piezoelectric sensors are used to identify the knee movement and facilitate the interaction between the user and the lower prosthesis through the socket. Piezoelectric sensors have been used to provide another technique that may help in the characterization of the knee movement. In addition, it may be compared to the FSR sensors to illustrate the extent of which both of them may be practically useful for the lower limb's designer. Moreover, the captured signals from the sensor assist in the development of prosthetic knee, in terms of the control strategy during different schemes. The piezoelectric sensors in this study were also placed inside the socket wall with specially made cavity to securely attach the sensor while allowing the required piezoelectric sensor deflection (Figure 3). Basically, one of the advantages of using piezoelectric bimorph is that it does not require external power supply to operate as it is considered an active sensor. Moreover, it also can be used to harvest energy when mechanical stress is applied on the bimorph surfaces [19, 20]. Basically, it consists of two layers sandwiched by metal layer for more flexibility as shown in a bimorph configuration as in Figures 2(a) and 2(b). Bimorph sensor is one of the most widely used bender actuators in both academic studies and industrial applications [16]. When applying pressure to the surface an electrical charge appears. The amount of charge is transferred into measurable output voltage which is proportional to the amount of pressure. The piezoelectric bimorph has a good dynamic characteristics in terms of handling transient inputs; also it has a wide range of output voltage up to ±90 V as well as a bandwidth about 100 Hz [16]. In addition to, the bimorph layer has a bleed resistor that protects it from high transient voltages and mechanical shocks.

Three piezoelectric sensors were attached at specific positions [15] at the anterior distal, anterior proximal, and posterior sites of the socket in order to sense the knee movement at different phases. A third piezoelectric sensor was placed at the anterior site nearby the knee joint to collect better measurement about the joint movement [1]. Figure 3 shows the placement of the piezoelectric sensors at both anterior and posterior sites.

2.2. Subject Characteristics and Experiments. A 29-year-old male, 75 kg, of height 182 cm transfemoral amputee who had been using an above knee prosthesis for the past 10 years, was recruited for this study. An informed written consent was attained from the subject as approved by the ethics committee of University Malaya Medical Centre. Two separate experiments with the same procedure were performed for each sensor, that is, FSR and piezoelectric sensors. In the first experiment, FSR sensors were attached at the regions of the quadrilateral double socket based on the subject's anatomical muscle position. The quadrilateral double socket was selected as it was the type of socket he had been using thus ensuring no compensatory gait deviations of using a new socket type. The sensors' wires were carefully secured and lengthened to ensure that the participant's movement was not affected. The amputee was fitted with the instrumented socket and knee prosthesis and was requested to perform five repetitions each of complete gait cycle, stair ascent, and sit to stand movements. The subject performed the stance phase of the gait cycle, that is, heel strike, flat foot, and toe off, as shown in Figure 4 for 5 repetitions. The subject was then requested to

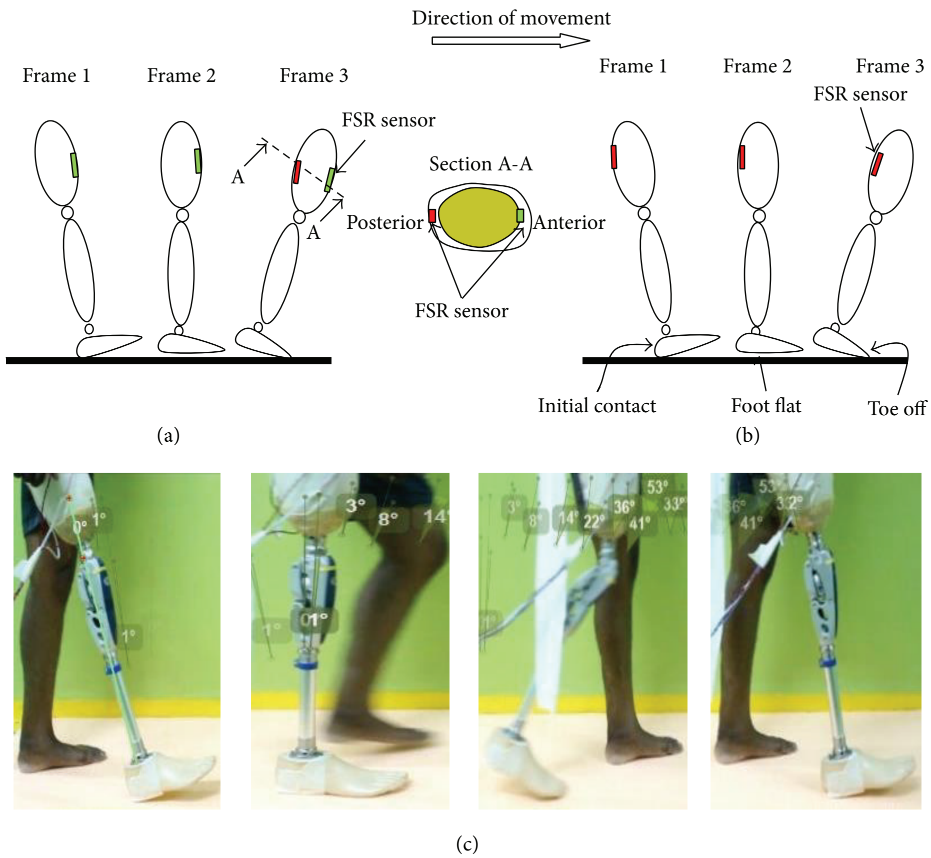

Figure 4: (a) Anterior FSR placement during full stance phase; (b) posterior FSR placement during full stance phase; and (c) the individual performing full stance phase (heel strike, flat foot, and toe off) while wearing the FSR instrumented socket.

go for stair ascent by positioning his leg in a flexed position upon an elevated step of 250 mm height (Figure 5), afterwards applying a downward force upon instruction. Finally, the amputee performed sit to stand action. The subject initially sat on a chair and stood up upon instruction (Figure 6).

2.3. Signal Processing and Movement Characterization. The signals generated from the user's activities were displayed and processed using Simulink (Real Time Windows Target Toolbox). The envelopes of the gait cycle curves were time aligned with the motion capture to define "heel strike," "flat foot," and "toe off" and processed to attain the amplitude patterns. The knee angle at each event was used as a reference to relate it with the captured signals as well as to show the ability of the sensors in characterizing the knee movement. Knee angle was captured by using Kinovea software and measured at each movement 30 Hz sampling rate in order to provide reference platform about the change during different phases. The curve profiles of the various movements were then characterized according to the standard deviation at specific points of each movement.

3. Results and Discussion

Variation of the captured signals versus time for FSR and piezoelectric sensors is presented in the following subsections. Knee angle was used as a reference for each case to relate the variation of the sensors output signals with the behavior of each knee movement phase.

3.1. Measurements of FSR and Piezoelectric Sensors throughout a Gait Cycle. This study protocol used FSR and piezoelectric sensors separately. A tethered FSR and piezoelectric sensors have been used. Using both sensors tethered together would add to the complexity of the setting which would inherently cause discomfort to the amputee subject thus producing unnatural gait.

The resulting FSR and piezoelectric sensors signals when performing different movements were compared. Figure 7 shows the FSR anterior and posterior outputs versus the knee angle throughout the gait cycle. The amplitude of both anterior and posterior sites started at heel strike. The pressure generated at anterior/posterior regions were the same as it produced output voltage of 3-3.1 V. However the knee angle at that phase is fully extended to begin the gait cycle. At about 33% of gait the FSR anterior output reached an amplitude of about 2.7 V. However voltage at the posterior sites remained higher than 3 V. At foot flat of 44% from the gait, the anterior voltage starts to increase and the posterior voltage has the same value of about 3 V. In addition, the knee angle started to flex before the time of foot flat preparing for the toe off stage. At the swing phase region which shows the maximum

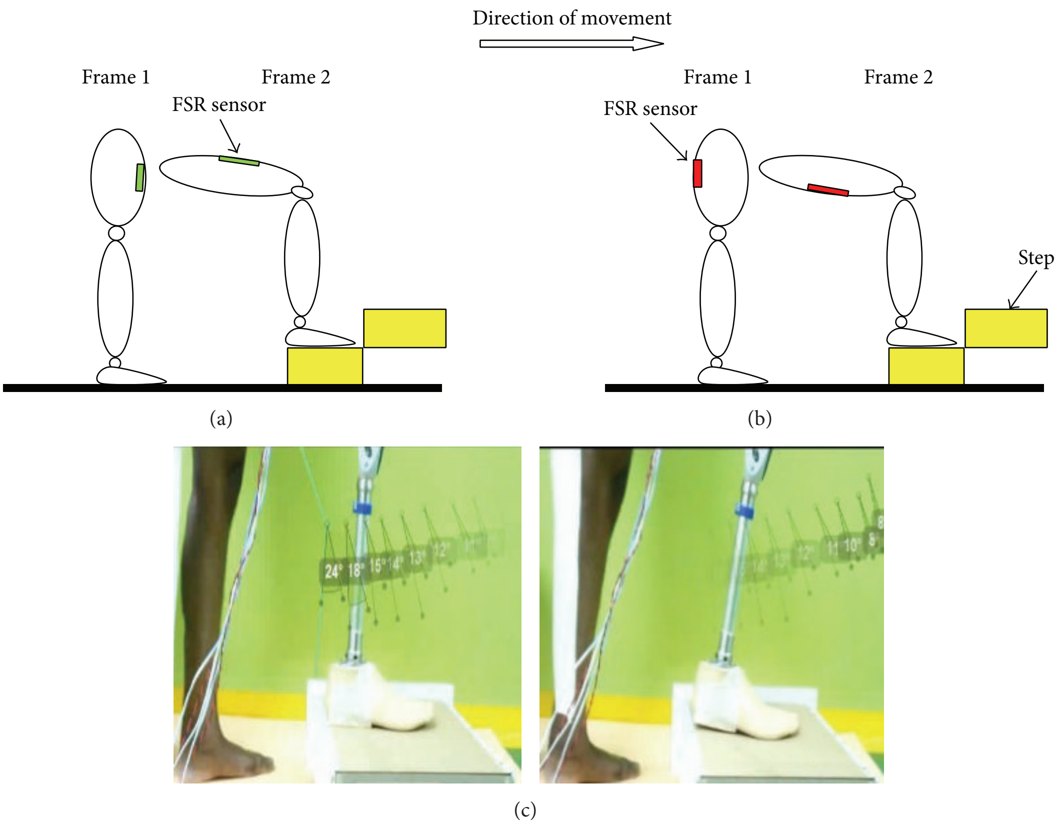

Figure 5: (a) Anterior FSR placement during stair ascent; (b) posterior FSR placement during stair ascent; and (c) the individual performing stair ascent while wearing the FSR instrumented socket.

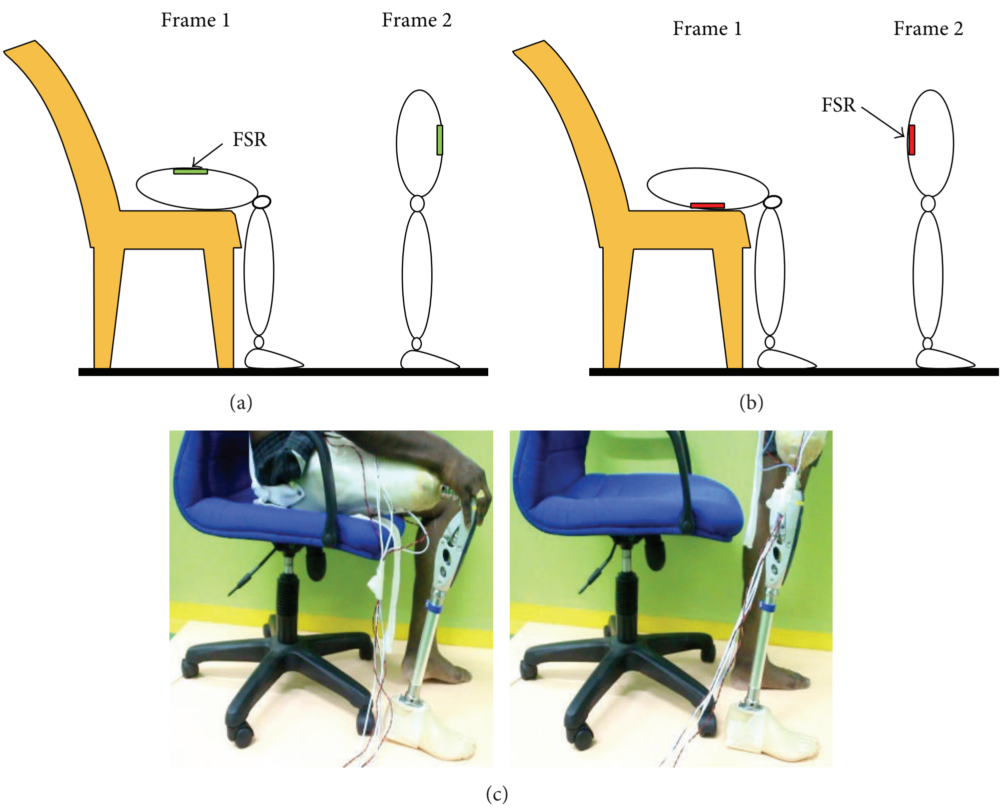

Figure 6: (a) Anterior FSR placement during sit to stand; (b) posterior FSR placement during sit to stand; and (c) the individual performing sit to stand while wearing the FSR instrumented socket.

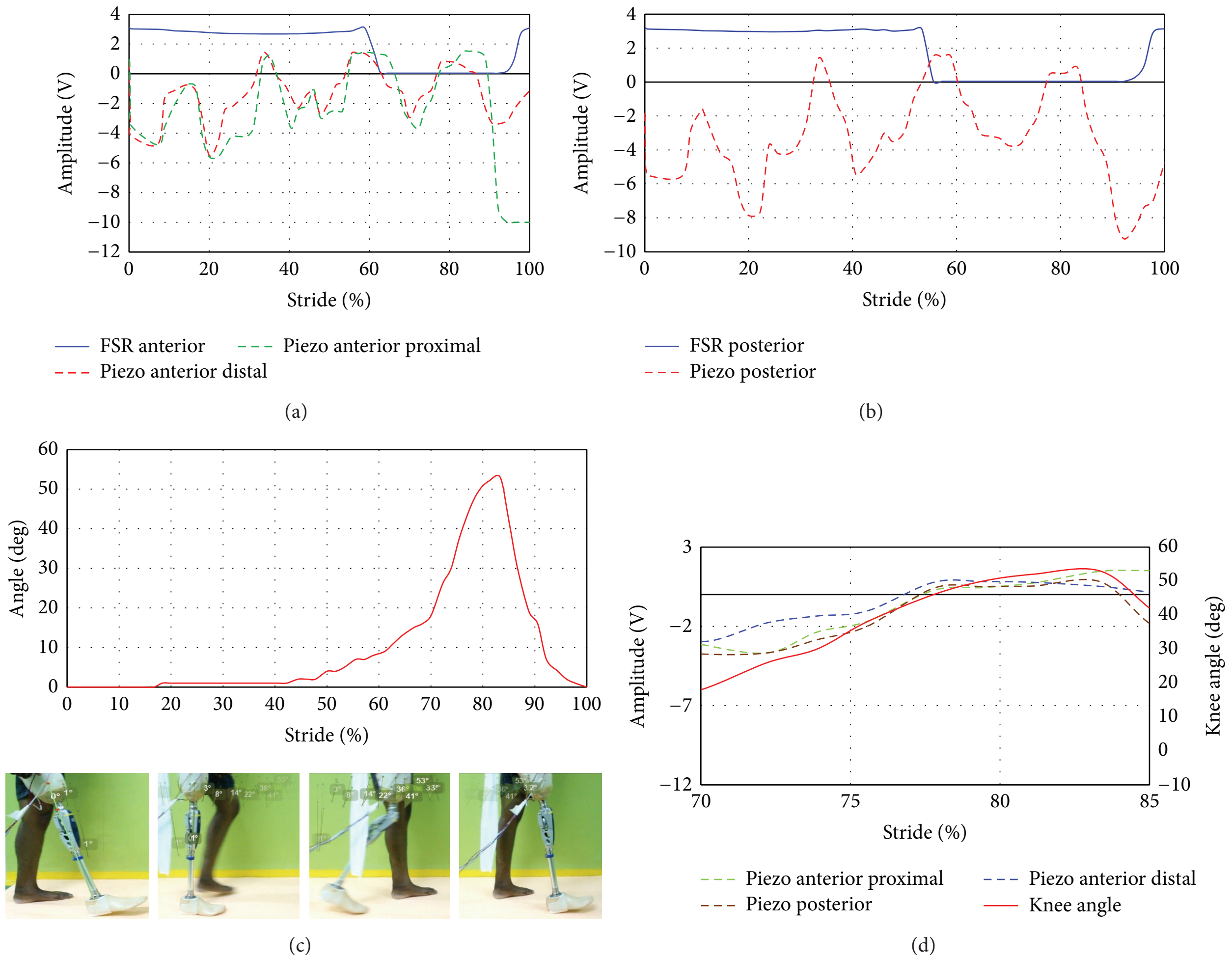

Figure 7: FSR and piezoelectric sensors output during gait cycle: (a) FSR anterior, piezo anterior distal, and piezo anterior proximal sites, (b) FSR and piezoelectric sensors at posterior sites, (c) knee angle during stride, and (d) piezoelectric sensors with the knee angle at a region from 70% to 85%.

knee flexion of about 53 degrees the FSR output of both anterior/posterior sites displayed minimum of about zero reading which indicates that there is no loading at both sensors at this stage. The gait cycle ended by reaching the full extension of the knee angle and increased the amplitude of anterior/posterior sensors up to 3 V. In essence, FSR could provide information about the gait change from the stance phase to the swing phase as can been seen from anterior/posterior graphs with the knee angle.

Results corresponding to the piezoelectric tests are conducted to be compared with FSRs' trials. Figures 7(a) and 7(b) showed that both anterior distal and anterior proximal sensors have the same trend line at 0–0.4 s of about 0–40% stride. The peaks of piezoelectric sensors demonstrated how the piezoelectric contracted once the pressure was exerted (positive peaks) and released when the piezoelectric retracts (negative peaks). As can be noticed from the knee angle lines during the swing phase at 1.6 s about 70% stride, the trend of both anterior proximal and posterior sensors matches the knee angle; moreover the posterior sensor exhibits similar behavior with the knee angle until the time reached 2 s. The behavior of the posterior piezoelectric sensor mostly had the same trend compared to the knee angle particularly at the swing phase. The toe off stage occurred at about 74% of the gait cycle, while the output voltage from the piezoelectric sensors intersected with neutral at zero voltage. This is because the generated pressure at this phase decreases due to unloading of the subject's leg from the ground. At the end of the gait cycle the output voltage became 10 V and 9 V at anterior proximal and posterior sites, respectively. Figure 7(d) illustrates the knee angle and piezoelectric sensors signals in the same graph. As illustrated in Figure 7(d), the trend of the piezoelectric sensor at swing phase (75%–85%) matches the knee angle behavior and the peaks cross the zero to the positive region. Figure 7(d) shows a closer look at the swing phase region from 70% to 85% to show agreement between the knee angle and the piezoelectric sensors.

3.2. Measurements of FSR and Piezoelectric Sensors during Sit to Stand. Similarly, FSR and piezoelectric sensors were used to measure the dynamic variation inside the socket during sit to stand movement. Figure 8 illustrated the FSR output versus

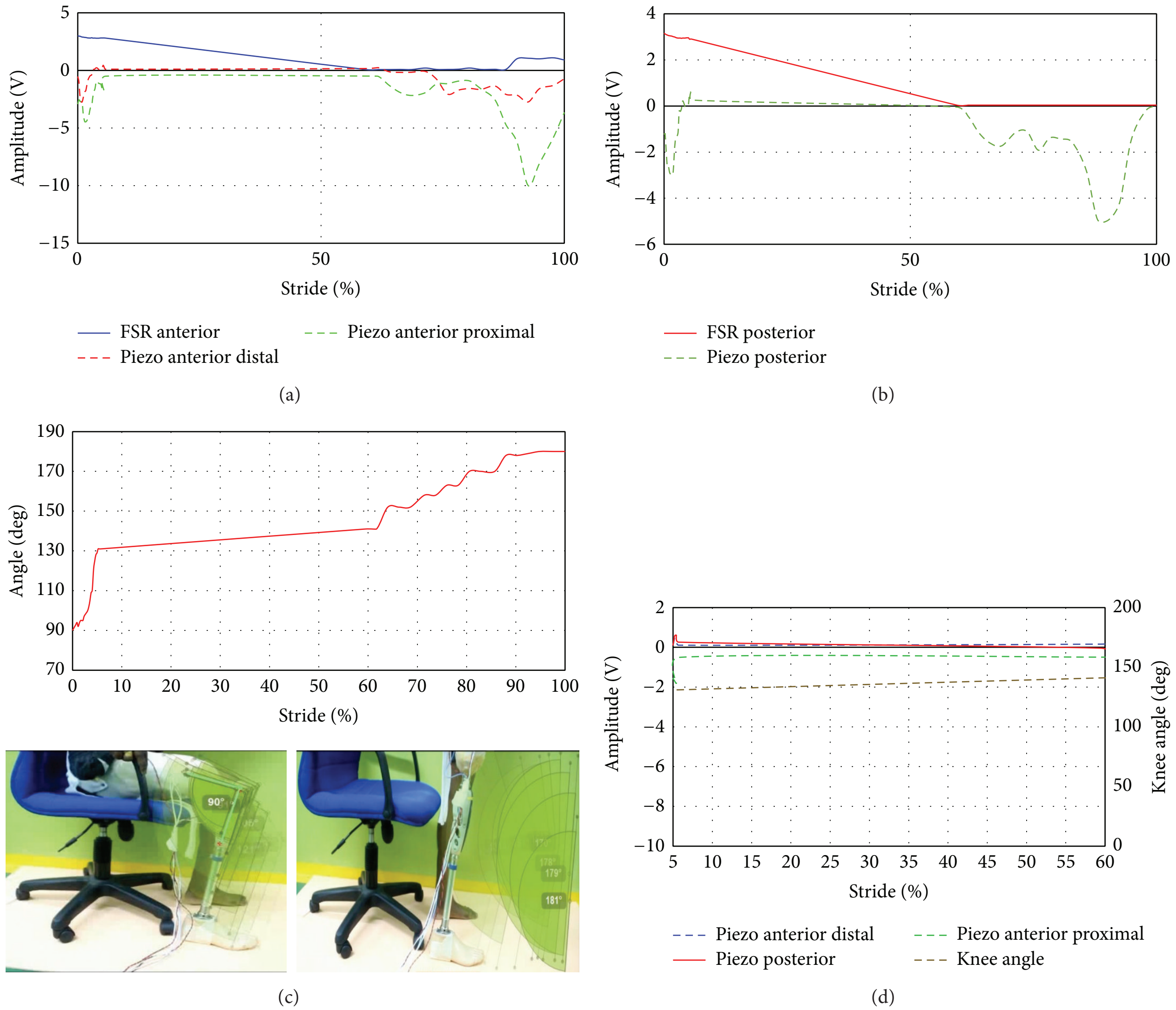

Figure 8: FSR and piezoelectric sensors output during sit to stand movement: (a) FSR anterior, piezo anterior distal, and piezo anterior proximal sites, (b) FSR and piezoelectric sensors at posterior sites, (c) knee angle during stride, and (d) piezoelectric sensors with the knee angle at a region from 5% to 60%.

the complete stride during sit to stand. The knee angle shown as a reference (Figure 8(c)) at the start of the sitting position was about 90 degrees opposite to amplitude of 3 to 3.1 V from both anterior and posterior FSR. The knee angle increased to 130 degrees at 5% of the movement. However the output of FSRs decreased below the 3 V, due to the pressure decrease at both anterior/posterior sites compared to the sitting position. The knee angle increased gradually to 180 degrees and consequently the anterior/posterior FSR sensors decreased linearly to the minimum value of about 0 V. Linear decrease of the FSR can be interpreted due to the sudden change of the movement by the subject which started from the sitting position to about 60% of the full stride before the full standing. This is one of the limitations of the FSR during that movement that should be considered in the future applications.

Sit to stand movement was tested and piezoelectric measurements versus stride were presented in Figure 8. The output signals from both anterior distal and posterior meets up from 50% to 60% have a zero voltage value, while at 60% to 100% of the stride, the piezoelectric sensor started to be decompressed as the voltage indicates negative value at that region. At anterior and posterior sites, two peaks of about 10 V and 5 V, respectively, can be noticed before the full standing position of the subject. As can be seen in Figure 8(d), a specific region from 5% to 60% was studied to show the relation between the knee angle and the piezoelectric signals. It is clear that the four signals of sensors and knee angle are straight line of about zero voltage for piezoelectric sensors and linear line of angle of a value of 140 degrees.

3.3. Measurements of FSR and Piezoelectric Sensors during Stair Ascent. Stair ascending was carried out as shown in Figure 9. The foot was placed on the step as shown in Figure 9 before the measurement of knee angle and sensors

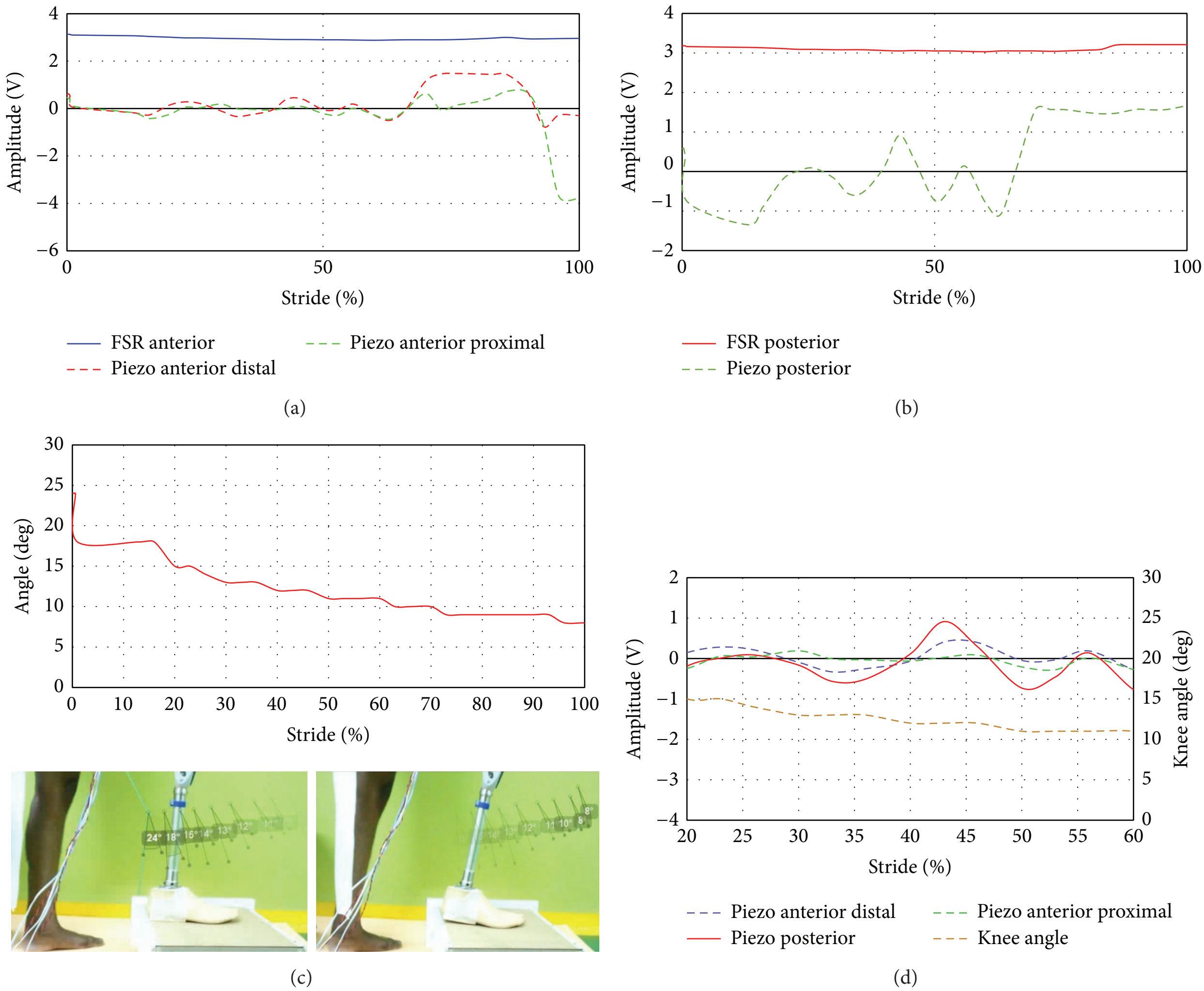

FIGURE 9: FSR and piezoelectric sensors output during stair ascent: (a) FSR anterior, piezo anterior distal, and piezo anterior proximal sites, (b) FSR and piezoelectric sensors at posterior sites, (c) knee angle during stride, and (d) piezoelectric sensors with the knee angle at a region from 20% to 60%.

is started. As illustrated in the graph, the output voltage of both anterior and posterior sensors remains almost constant during the whole event because of the pressure generated from the ground, which is directly reflected as voltage of about 3–3.2 V. The knee angle varied from 23° to 9° at the end of the stair ascent phase. Stair ascent movement was conducted with the user wearing the socket embedded with the piezoelectric sensors. The knee angle decreases gradually from about 23° to 8°; however the variation of the output signals from piezoelectric sensor at both anterior distal and posterior proximal sensors changed minimally during the 0% to 60% stride. Piezoelectric sensor at the posterior site decompressed at the early stage of the stride at 10%. High compression value was noticed at anterior distal site which has a value of about 1.5 V (Figure 9). In overall, Figure 9(d) shows the three piezoelectric signals with the knee angle in the same graph. It can be noticed that the fluctuations of the piezoelectric sensors agreed at a region starting from 20% to 60%. This region can provide information when compared with the variation of the knee angle which starts from 15° to almost 10°.

Analysis was conducted to identify the events during the gait cycle based on the events as described by Nordin and Frankel [22]. The swing phase is divided into initial swing (60–73% of gait cycle), midswing (73–87% of gait cycle), and terminal swing (87–100% of gait cycle). FSR output signals showed some delay during the transition from stand to swing as a result of the FSR characteristics reported that it has 1-2 ms mechanical rise time delay [17]. Therefore, at the walking phase the results of FSR are considered with the mentioned delay and piezoelectric sensors can function better than FSR. Results of the piezoelectric sensors (Figure 8(d)) can be combined to describe midswing and terminal swing events. Figure 9(d) illustrates good agreement between knee angle and the piezoelectric sensors within a range of voltage from −4 V to −2 V and the knee angle proportionally changed from 20° to 55°. Sit to stand phase is important to the transfemoral amputees and the movement events can be identified from

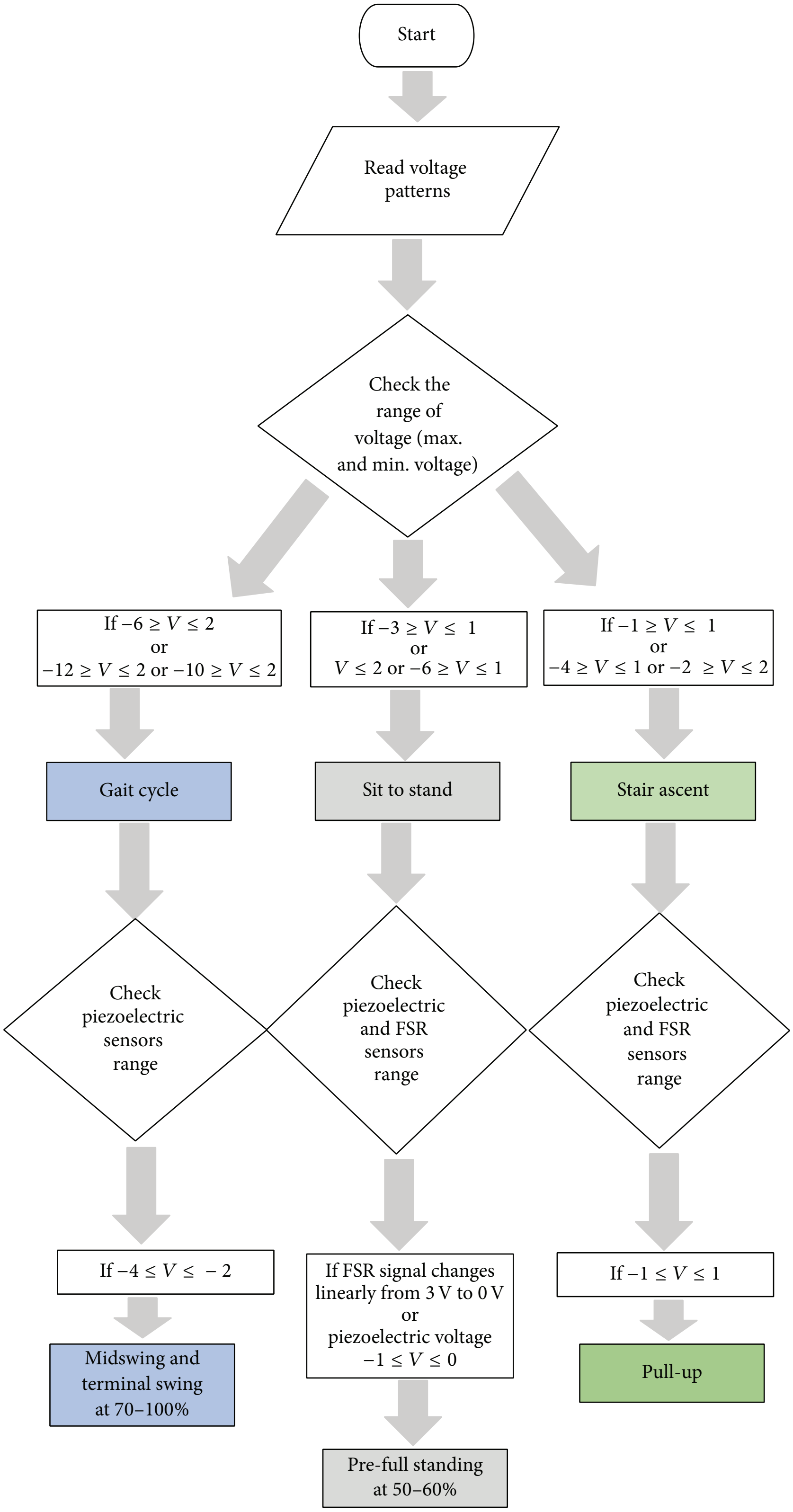

Figure 10: Flow chart presents the identification process of the knee state by consideration the range of voltage of piezoelectric sensor.

Table 2: Standard deviation values for FSR.

	Gait cycle %	Standard deviation ±	Sit to stand %	Standard deviation ±	Stair ascent %	Standard deviation ±
Anterior	0	0.016	0	0.032	0	0.058
	0.17	0.026	0.23	0.041	0.3	0.074
	44.44	0.047	0.5	0.046	0.6	0.099
	46.11	0.064	59.4	0.029	30	0.165
	62.78	0.000	61.7	0.027	33	0.173
	64.44	0.000	64	0.025	36	0.180
	66.67	0.000	66.3	0.029	40	0.189
	70	0.000	97.7	0.028	96	0.159
	72	0.000	100	0.024	100	0.151
	97.78	0.047				
	100	0.018				
Posterior	0	0.025	0	0.020	0	0.005
	0.17	0.027	0.23	0.010	0.3	0.015
	44.44	0.0 ± 04	0.5	0.013	0.6	0.029
	46.11	0.124	59.4	0.075	30	0.012
	62.78	0.000	61.7	0.078	33	0.014
	64.44	0.000	64	0.081	36	0.015
	66.67	0.000	66.3	0.082	40	0.000
	70	0.000	97.7	0.030	96	0.017
	72	0.000	100	0.027	100	0.019
	97.78	0.004				
	100	0.019				

the signal pattern. The prestanding phase at 50 to 60% of the movement can be recognized from both FSR and piezoelectric signals (Figure 9).

Stair ascent movement was divided into five submovements [23]. The pull-up submovement could be determined by considering the piezoelectric signals while its voltage was between −1 and 1 V (Figure 9). Flowchart shown in Figure 10 concludes how the results conducted from the current study are used to build an algorithm to identify specific events during different knee movement. Walking gait, sit to stand, and stair ascent can be identified according to the flow chart. Specifically, midswing and terminal swing can be recognized. At sit to stand movement, pre-full standing event can be seen at 50–60% of the stride. Finally, pull-up event can be identified at the stair ascent movement. The variation of both FSR and piezoelectric sensors readings at specific points during each movement was reflected as the standard deviations in Appendix Tables 2 and 3. As can be noticed the wide range of measurements of piezoelectric sensor will help to identify the knee movement.

4. Study Limitation

This study was performed to establish the proof of concept with a single amputee subject particularly to look at the different sensor responses. The session was conducted with five trials per movement represented by the standard deviation values at the Appendix section. To ensure natural walking, quadrilateral socket was used in this study as it is the type of socket that is used by the subject in his daily activities. It was also assumed that the middle of the muscle belly is the area of greatest pressure within the socket, and in this case study it was verified by the greatest pressure felt during the subject's maximum voluntary contraction through manual palpitation of the muscles. In other cases, it could depend largely on the socket fit; thus this factor should be taken into consideration in further studies. Additionally, the current study indicated that the piezoelectric sensors could be useful in recognizing the knee movement better than the FSR because of the variations shown during each phase. More experiments should be conducted with different socket types in order to make better comparison between both sensors used in the current study. Moreover, statistical significance can be obtained by considering more than one subject to make the results more convincing.

5. Conclusion

This study presented the possibility of identifying the submovement of a transfemoral amputee using FSR and piezoelectric sensors integrated into the socket. A pair of FSR and three piezoelectric sensors were embedded separately at anterior and posterior sites inside of the socket to be directly in contact with the residual limb of a transfemoral amputee.

Table 3: Standard deviation values for piezoelectric.

	Gait cycle %	Standard deviation ±	Sit to stand %	Standard deviation ±	Stair ascent %	Standard deviation ±
Anterior distal	0	0.448	0	0.783	0	1.276
	0.16	1.046	0.22	1.123	0.3	1.102
	42.22	3.057	0.45	3.707	0.6	1.202
	44.44	3.278	5.71	0.252	40	0.138
	46.11	4.086	59.42	0.422	43	0.006
	72.22	2.873	61.71	0.892	46	0.207
	73.88	0.192	64	1.203	50	0.203
	97.77	0.963	66.28	1.694	53	0.224
	100	0.811	97.71	2.149	96	1.308
			100	2.346	100	1.336
Anterior proximal	0	0.849	0	4.617	0	0.654
	0.16	0.712	0.22	3.616	0.3	0.703
	42.22	4.909	0.45	2.428	0.6	0.705
	44.44	3.975	5.71	2.264	40	0.603
	46.11	6.115	59.42	3.961	43	0.044
	72.22	6.031	61.71	4.918	46	0.117
	73.88	5.598	64	5.075	50	0.479
	97.77	1.096	66.28	3.194	53	0.476
	100	1.173	97.71	1.853	96	0.497
			100	1.833	100	0.598
Posterior	0	2.375	0	2.026	0	1.439
	0.16	5.215	0.22	1.650	0.3	1.129
	42.22	5.284	0.45	3.168	0.6	1.442
	44.44	6.136	5.71	2.639	40	0.141
	46.11	5.715	59.42	3.643	43	0.189
	72.22	2.853	61.71	4.032	46	1.253
	73.88	1.628	64	3.375	50	1.270
	97.77	6.081	66.28	3.119	53	1.308
	100	6.048	97.71	4.917	96	1.654
			100	4.930	100	1.675

Complete gait cycles as well as stair ascent and sit to stand motions were performed by the transfemoral amputee to determine the predictability of the knee movement detection as well as user intention by using FSR and piezoelectric sensors. This would be useful in further studies related to the prosthetic knee development. The piezoelectric sensors indicated wide range of measurements at all conducted movements. In particular, piezoelectric sensors can identify submovements at gait and stair ascent movements within a specific range of output voltages. In addition, signals from piezoelectric sensors show acceptable agreement while tracking the knee angle at gait cycle and sit to stand. However, more work should be considered for using piezoelectric sensors at stair ascent/descent and slope climbing. In case of FSR, it could be useful in detecting the change of gait from stance phase to swing phase. FSR showed that it could be used in identifying the pre-full standing phase at sit to stand movement. Therefore, one of the recommendations from this study is that FSR may be more useful to be used as a trigger between the knee movements (walking, sit to stand, and stair ascent) due to its measurement limitations and would complement the piezoelectric signal for major movement detection.

Following this efficacy study, it can be concluded that the user's intended movement could be detected prior to its angular mechanical change using an instrumented socket. Further trials are to be conducted with greater sample size to determine the consistency and accuracy of response in different subjects with different residual limb lengths, socket types, and muscle condition. This study also demonstrated that piezoelectric sensors could be safely and effectively be embedded onto the socket wall to provide reliable response signal that may be helpful in recognizing the user intention and maintain the amputee's comfort and normal stride while wearing his prosthesis. However, more subjects and simulation of different sensing methods are recommended to address more variations in sensor responses. The proposed approach presented in this study could serve as a complementary input to optimize the interaction of the user with the existing or new microcontrolled prosthetic devices.

Appendix

See Tables 2 and 3.

Conflict of Interests

The authors declare no conflict of interests.

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Modelling and Control of a Linear Actuated Transfemoral Knee Joint in Basic Daily Movements

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Abstract: This paper presents the modelling and control of an actuated prosthetic knee mechanism for transfemoral amputees. The mechanism consists of a linear actuation system that feeds the mechanism with the required moment and power at different movements. Physical simulation was utilized to simulate and identify the all physical parameters of the knee. Particular pattern of the knee angle was used as a reference to test the behavior of the knee mechanism in terms of the angle and speed. Movements such as sit- to- stand, slope climbing, and stair ascent were tested at different time intervals. PID control parameters were tuned while the angle of the actuated knee mechanism could track the desired angle at time period of 1 s and 0.1 s at different movements. However the mechanism showed deviation from the desired input at time periods of 0.05 s and 0.0125 s. In addition, the estimated amount of torque and power at time period less than 0.1s were about 15 N and 800 W. The physical simulation presented a realistic simulation of the actuated mechanism in terms of the knee parameters. Further analysis may be carried out during the development stage of the knee mechanism. Also, more experiments could be conducted with the transfemoral amputees to improve the overall performance of the knee mechanism.

Keywords: Linear actuated knee mechanism, Physical modelling

1 Introduction

The human knee is a complex structure that could perform different activities and movements. For above knee amputees, powered lower limb prosthesis are still in the development stage. As the transfemoral amputees are in need of performing several daily movements like other people, thus the progress in development of prosthetic knee mechanism is essential. Currently, some powered lower limb could mimic the normal walking activity of the transfemoral amputees [1,2,3]. The powered lower limb consists of knee and ankle prosthesis, and it could deliver sufficient power to assist the amputee during walking. Also, the powered limb system could change the impedance during the activity and it was verified through experimental studies with an amputee patient.

Another development is the prosthetic foot that uses spring that could store and release energy during different phases [4]. The active knee prosthesis has been developed using different types of actuator that could provide sufficient torque to move the knee joint. For example, a magnetorheological (MR) fluid actuator was utilized to develop the prosthetic knee joint, as MR was used in a shear mode to effectively modify the required torque. Also, pneumatic and hydraulic systems are used to change the damping of the knee mechanism during movement [5,6,7]. Pneumatic muscles were used as an approach to develop a prosthetic knee which behaves similar to the biological knee, and results revealed that the system can replicate the normal gait cycle [8]. So far, it can be addressed that the experimental studies that investigate the level ground walking were presented. However, similar studies presenting different activities such as stair ascent, stair climbing, and sit- to- stand are still very scarce. A preliminary validation of using a powered lower limb prosthesis during stair ascent and descent was presented and clinically evaluated with a transfemoral amputee subject [9]. The experiments showed appropriate performance related to the joint kinematics during the stair ascent trials. Standing movement was evaluated and a control scheme was developed to control the powered prosthesis during standing movement [10]. The results indicated that the

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controller could assist to deliver positive power during standing up.

It is anticipated from the prosthetic knee mechanism to rotate within a certain range to perform various movements such as walking, stair ascent, sit- to- stand, and slope climbing [11,12]. In most activities, the maximum flexion of a normal knee angle is about 140°, but practically the knee does not fully bent up to 140°. The flexion angle range required to accomplish the activities range only from 0°- 120° [13]. It is requested from the amputees to be able to perform most of daily activities by allowing the prosthetic knee to flex up to 120°. Therefore, one of the goal of current is to make the knee mechanism flexes within a range of 0°- 120° to meet the basic daily activities. In order to accomplish that range of angle, the knee mechanism should articulate using an actuation system that could deliver a sufficient amount of torque to perform that task.

In particular the knee joint that could perform various daily activities is dependent on the control of the actuation system. In other words, the actuation system is responsible of generatintg sufficient force and varying the impedance might enhance the development of the prosthetic knee mechanism [14] by adopting a Mechatronic approach actuation system can be designed and tested during the that operation of the knee mechanism at different level of movements. The Mechatronics approach offers an opportunity to utilize a pre-design investigation of the system using a physical modelling. That physical modelling is essential before the actual development, as it could provide a sufficient information about the system before the actual physical development stage [15]. In addition, the physical modelling tool provides a comprehensive view about the overall system in terms of its dynamic behavior rather than using the conventional modelling method of deriving the dynamic equations [16]. Also, the physical modelling can perform a complete simulation of the entire system and assists to update the system components and parameters during the stage of the design.

Different structures of the lower limb prosthesis were introduced in previous studies. In this study, the main demand is to control and evaluate the performance of the actuated knee mechanism and adjust the actuation system parameters at various movements. In other words, the assessment of the performance of the knee mechanism and the actuation system in terms of the overall parameters shall provide a clear platform for the development of the prosthetic knee mechanism. In this study, stages of developing a prototype of prosthetic knee mechanism as well as the physical modelling of the updated prototype of the knee mechanism were presented.

In this study the lower limb at normal ground walking was physically simulated. An estimation of the knee joint damping and spring stiffness were estimated for the purpose of validating the concept of setting the actuation system of the knee mechanism. An actuated knee mechanism that utilized a motor-spindle system was modeled using Simulink™ developed by MathWorks®. The desired angle patterns were fed into the actuated knee mechanism in order to test the performance of the knee joints inclination at various knee movements of walking, sit-to-stand, stair ascent, and slope climbing. Finally, the overall parameters of the knee mechanism and actuation system in terms of the damping coefficient and spring stiffness were concluded. Moreover, the main features of the actuated knee mechanism were listed.

2 Materials and Methods

2.1 Simulation of multi-body of the lower limb

In order to start the development of an actuated knee joint, it is recommended at the first stage of the process to simulate the lower limb movement for the purpose of estimating the dynamics of the lower leg. The simulation of the lower leg can be performed using a physical modelling (multi-body simulation) using Simulink™ developed by MathWorks®. Basically, multi-body systems are used to model the dynamic behavior of interconnected rigid bodies that have their relative motion constrained by kinematic joints and that are actuated by forces or moments. The multi-body model was developed in the current stage and presented in Figure 1. The data was obtained from, a person weighting 56.7 kg has a corresponding leg mass of 2.63 kg [17]. The lower leg is composed of the foot (body), ankle (joint), leg (body), and knee (joint). Each body element was characterized by mass, moment of inertia tensor, center of mass, and dimensions, while the joints were defined by the degree of freedom (DOF) and constraints.

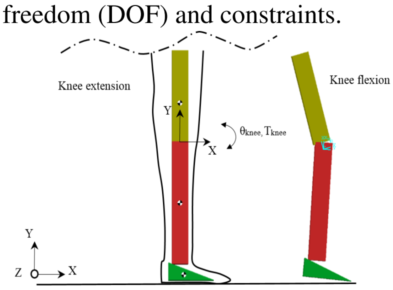


Figure 1. Scheme for the multibody knee model with the representation of kinematics and kinetics prescribed at the knee joint.

To carry out the model analysis at the first stage of the development, it is essential to determine the kinematics of each joint. Translational movement occurs at (X,Y) coordinates, while rotational movement occurs around the Z axis, at the sagittal plane. Basically, translational and rotational movement was determined relative to the knee joint, where the leg, ankle joint, and foot elements were connected, respectively. The rotational movement of the leg through the knee joint was prescribed relatively to the position of the leg, where the neutral position of zero degree of the leg occurs when the leg and thigh are totally extended. According to the joints kinematics, it is necessary to set the parameters of the knee joint model to use. In order to evaluate the design of the prosthesis, a prosthetic knee model is proposed (Figure 2). Spring damper arrangement is proposed to replicate the knee joint behavior. As a first iteration of the design, control algorithm was built to control the angle of the knee joint with reference to a pre-defined normal walking pattern [18]. Variation of the lower limb joint angles during the stride was simulated and presented in Figure 3.

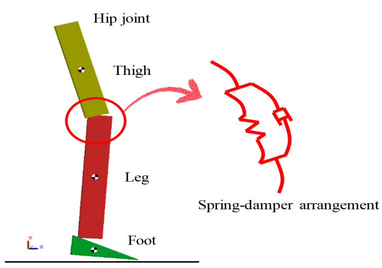


Figure 2. Multibody biomechanical model used in the simulation with a simplified spring-damper arrangements.

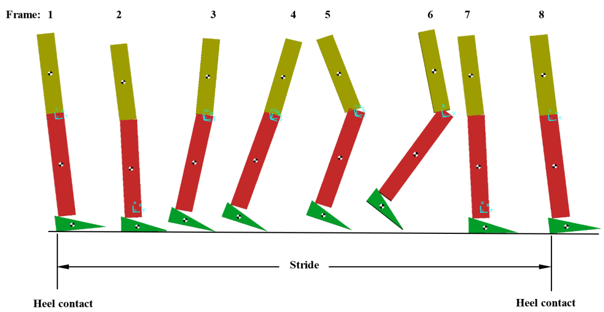


Figure 3. Lower limb different scripts during stride.

To mimic the normal knee angle movement and set the parameters of the knee joint during the stride, different modes of control algorithm were implemented (P, PI, PD, and PID). The lower leg range of motion measured in degrees was established as close as possible to the leg human range motion during level-ground walking by applying PID controller [19]. The damping coefficient and spring stiffness of the knee joint were listed in Table 1 along with the controller parameters. With all joints parameters prescribed, it was necessary to set the parameters of the joint model to be used as a platform at the next stage of the development.

Table 1. Parameters of the knee joint during stride.

Variable	Value
Spring stiffness	0.1 N.m/deg
Damping coefficient	0.3 N.m/(deg/sec)
Proportional gain	25
Integral gain	47
Derivative gain	0.14

2.2 Stages of developing the actuated prosthetic knee mechanism

The assistance of the transfemoral amputee to replicate different gait phases can be provided by a prosthetic knee device [20,21]. As described in the prior section the multibody modelling of the lower limb was implemented. Inhere, detailed information about the proposed actuated knee mechanism is discussed. In order to move the prosthetic knee mechanism, the actuation system must produce enough power to provide the knee movement during different phases.

2.2.1 Mechanical design and development stages of the actuated knee mechanism

The development of the actuated prosthetic knee has gone through several stages and ideas based on the Mechatronics approach [22]. First prototype of a prosthetic knee system was developed by the authors, which consists of knee mechanism components that were fabricated using aluminium 6061. The knee joint consists of a simple, hinged based structure. It allowed movement of a single degree of freedom in the sagittal plane. The actuator was a servomotor (Maxon® EC 32, brushless) that was connected to spindle drive to supply the sufficient force to the knee joint. The motor was capable of operating at 9460 rpm constant speed. A spindle drive GP 32 S from Maxon® was connected into the motor in order to increase the output torque a well as decrease the motor speed. Rotational motor torque was transmitted by the metric spindle screw to linear output force via a lead screw assembly. Linear force was transmitted to angle-dependent rotational torque about the knee joint via the moment of arm. The motor spindle drive was mounted to the mechanical structure of the knee joint to achieve $0°$- $60°$ range of motion, which was the required range of motion for walking and sit-to-stand.. However, the system could not meet the speed requirement.Therefore, the update of the early version was necessary to improve such movement performance.

Based on the limitation of the first prototype, a new design was adopted and tested. The proposee design should be able to perform movements such as walking, stair ascent, sit- to- stand, and slope climbing. The actuated prosthetic knee mechanism was physically modeled in order to move the knee at a range of motion from 0°- 120°. The design was implemented using CAD software, SolidWorks®. The physical simulation of the actuated prosthetic knee was performed using Simulink™ MathWorks®. As shown in Figure 4, the mechanism is similar to the crank slider mechanism cite1. Different isometric view of the actuated prosthetic knee is shown in Figure 5. As the knee joint has to rotate to achieve walking, sit- to- stand, stair ascent, and slope climbing movements. The actuation system is responsible for delivering the required output mechanical energy to move the knee joint. Thus, actuation system parameters should be tuned to meet these requirements at each movement. Therefore, the spring stiffness and damping coefficient at each phase should be determined accordingly.

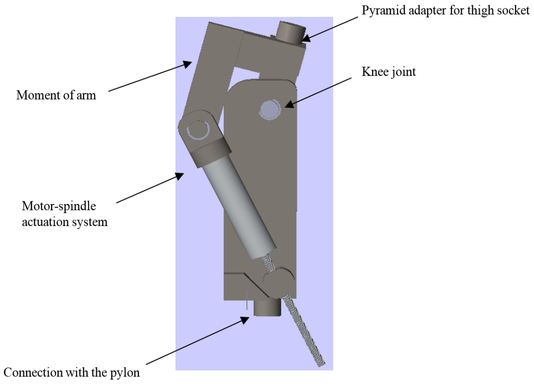


Figure 4. Overall view of the proposed motor- actuated prosthetic knee.

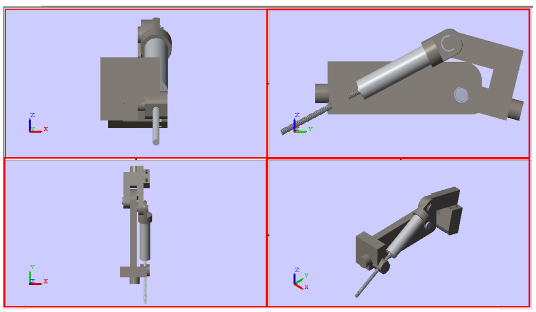


Figure 5. Different isometric views of the actuated prosthetic knee.

2.2.2 Physical modelling of the actuated prosthetic knee mechanism

The main focus of the current study is to develop the actuation system of the knee joint that could assist to perform different activities. Thus, a physical modelling of the knee joint mechanism was modeled using Simulink™ MathWorks ®. The advantages of such method of physical modeling is to simulate and mimic the actual system by adjusting all parameters and variables. As can be seen in Figure 6, the overall block diagram of the knee joint was physically modeled and the details components of the knee components were linked and simulated as in Figure 7. As discussed earlier during the lower leg simulation, the estimation of the spring stiffness and damping coefficient are useful to select an appropriate actuation parameters for each knee movements. The actuation system used to actuate the knee mechanism is motor- spindle drive. The motor- spindle drive produces the required force to move a prismatic joint. At the end, the force is converted to the knee torque by means of the moment of arm. In order to control the actuation system, it is recommended to get the simple model of the knee joint.

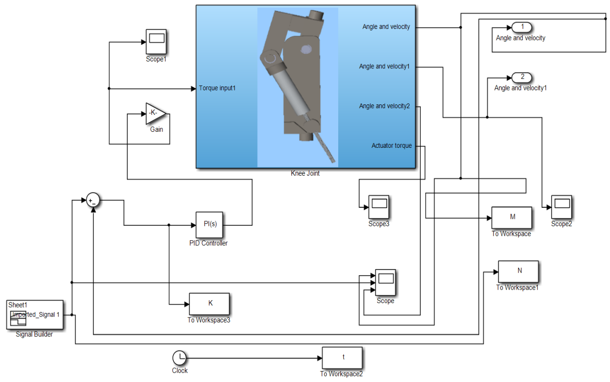


Figure 6. Overall physical block diagram of the prosthetic knee joint.

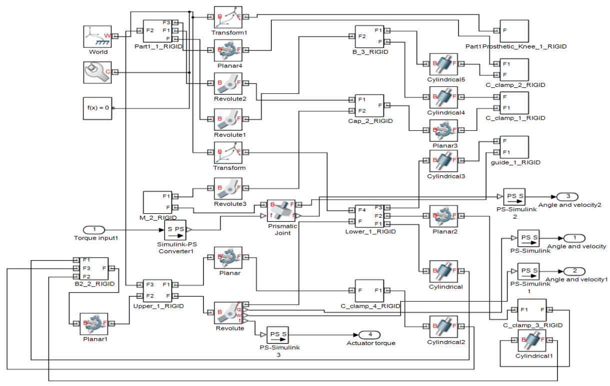


Figure 7. Detailed physical modelling of the knee joint mechanism including the all the components and joints.

The simplified knee joint model can be obtained Newton second law for rotational motion as in equation (1) [23].

$$\tau_k = J_k \theta^{\cdot\cdot} \tag{1}$$

where J_k is the knee rotational inertia due to foot mass, and $\theta^{\cdot\cdot}$ is the knee angular acceleration. Equation (1) can be written as a transfer function using the Laplace transformation, resulting equation (2).

$$G_k(s) = \frac{\theta_k(s)}{\tau_k} \tag{2}$$

The resulting transfer function represents a second order system which will be considered in the design of the control system of the actuation system along with the knee joint.

2.2.3 Performance of the actuation system and knee joint at different movement phases

During the development of the knee joint, the actuation system could assist the amputees to perform activities. Therefore, appropriate control strategies have to be implemented. It is necessary to devise a control scheme where the manipulated and controlled variables are to be adjusted. The control scheme presented in Figure 8, is a closed-loop control system, where the difference in the desired and actual condition creates a correction control command to remove the error [24]. The control scheme produces the a suitable signal to operate the actuation system that produces the required force, in addition the master function of the controller is to adjust the knee angle at different movement trajectories.

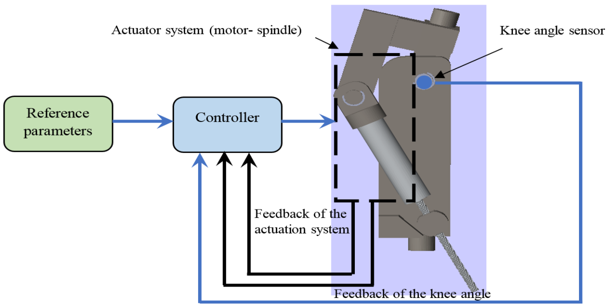


Figure 8. Diagram of a simplified control scheme for controlling the knee angle movement.

3 Results and Discussion

To evaluate the capability of the proposed actuated knee mechanism, different trajectories were chosen to test the mechanism. Walking, slope climbing, stair ascent, and sit-to- stand movements were adopted as a reference patterns. Afterwards, an appropriate control scheme was developed to track the reference trajectories of each movement. Also, a step response was used to test the behavior of the knee joint at specific knee flexion angles. Also, to check the performance of the actuation system, the force produced from the motor-spindle to actuate the knee mechanism was presented at each movement.

3.1 Testing the knee mechanism at specific flexion angles

The actuation system is chosen based on a suitable knowledge of the knee joint requirements that could move the knee joint at a specific range of motion. The force generated from the actuation system could move the knee mechanism at various knee flexion positions. The actuation system was controlled using the PID control scheme that was developed and a specific knee joint positions were tested at different flexion angles (Figure 9). As shown in equation (2), that the simplified model of the knee can be represented in a second order form. Thus, the PID control scheme was applied to check the performance during the different knee flexion. The actuation system shows a capability to move the knee joint within the specific range required from 0° - 120°.

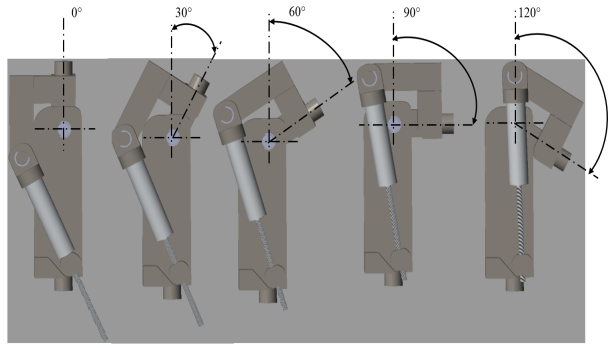
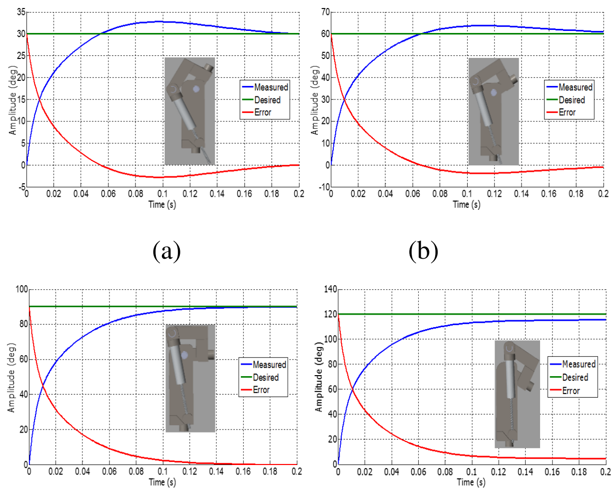


Figure 9. The prosthesis positions at the different knee flexion angles.

As the knee angle varies suddenly at certain phases during the stride. Subsequently, a step input response at different knee flexion shown in Figure 10 was conducted while using PID control scheme. It was assumed that the knee joint responses according to the desired input occurs at a rapid time, thus a 0.2 s was adopted as an interval to check the response [25]. As can be seen four states of the knee angles were tested and the responses were recorded and compared with desired input. The step responses at 30° and 60° show less oveshoot and settling time. In contrast, the 90° response exhibits smoothely without overshoot and shows rapid settling time. A steady state error at the 120° response is clear, as the controller attempts to settle the mecanism inertia in a small period of time.



(a) (b)

(c) (d)

Figure 10. PID control scheme responses at 30°, 60°, 90°, and 120°, (a), (b), (c), and (d).

3.2 Testing the knee mechanism with the normal knee angle profile at various speeds

In order to test the performance of the knee joint, a reference pattern of the normal walking was applied as an input to the knee system. For further analysis, the knee mechanism was tested under different time bases, 1, 0.1, 0.05, and 0.0125 s. the purpose of using different time base is to check the knee performance under various speeds (Figure 11 and 12). In addition, to test the performance of the mechanism dynamics at various conditions. The torque and power delivered from the actuation system to the knee mechanism can be calculated at different speeds. The output knee angles were obtained and compared to the input signal at PID control algorithm.

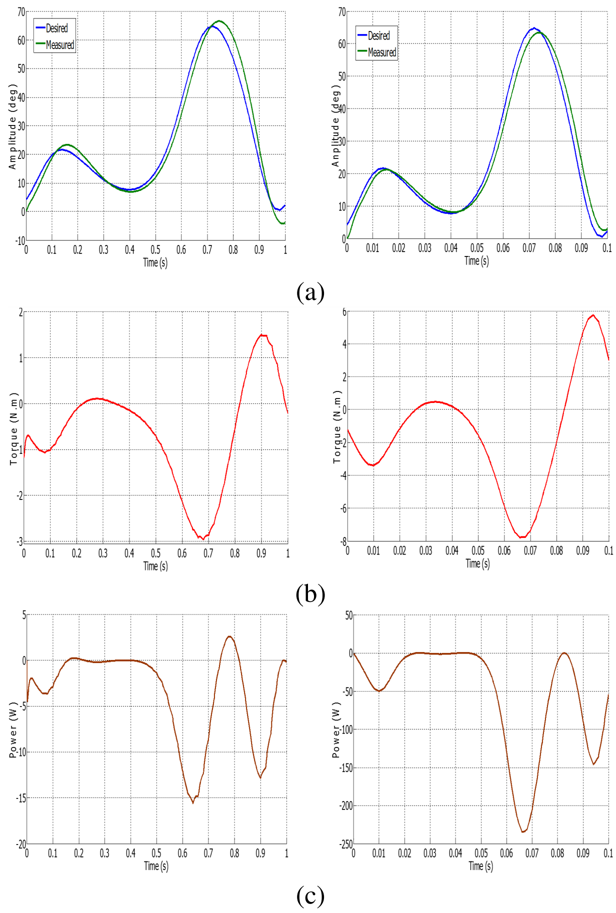


Figure 11. Knee angle, torque and power delivered from the actuator at different time bases, 1 s and 0.1 s, (a), (b), and (c)

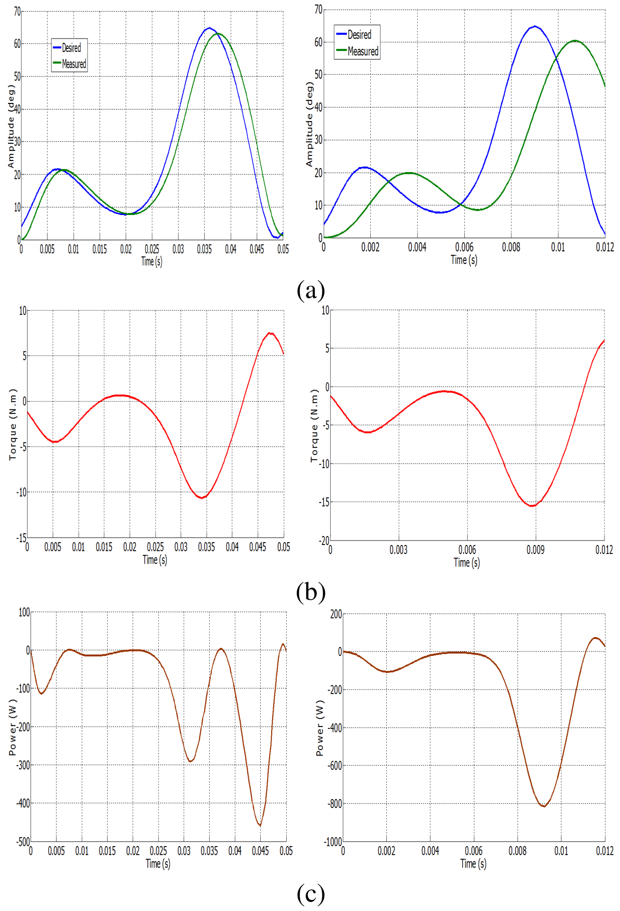


Figure 12. Knee angle, torque and power delivered from the actuator at different time bases, 0.05 s and 0.0125 s, (a), (b), and (c).

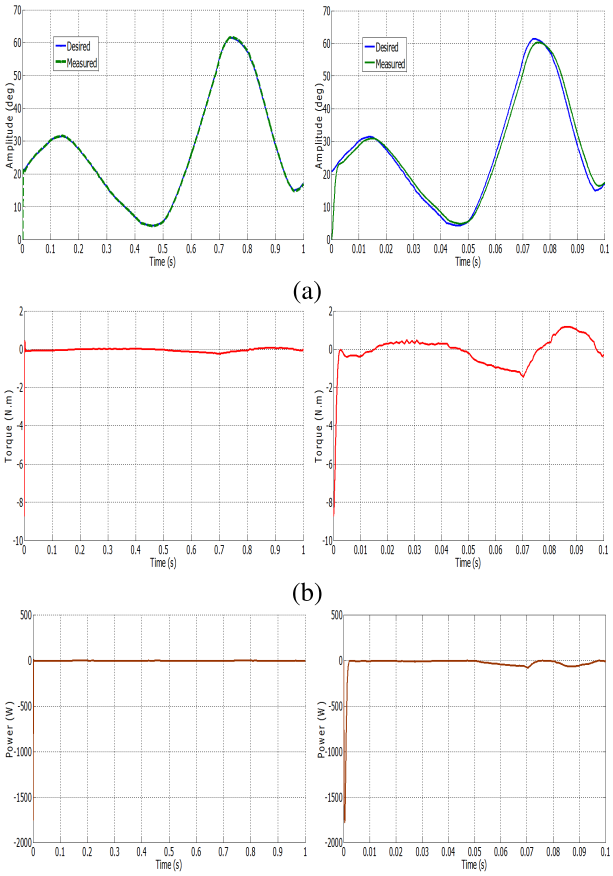
In general, Figure 11 shows the knee angle performance at 1 s and 0.1 s intervals, it can be observed that the knee mechanism could track the normal gait. The fluctuation of torque and power at 1 s and 0.1 s were recorded to show the change of torque and power rates during the time intervals. In contrast, the performance of the knee angle at less time intervals of 0.05 s and 0.0125 s indicate some delay from the system during tracking the desired angle profile (Figure 12). The output profile of the knee angle at 0.0125 s shows delay at the two peaks at stance and swing phase which could not track the desired input. Also it can be noticed that the amount of power and torque required to provide the movement are 15 N.m and 800 W respectively. In comparison with low speeds the torque and power rates are about 7 N.m and 200 W respectively. The performance of the current knee mechanism at time base less than 0.1 s shows large amount of torque and power that accommodate with the normal knee pattern.

3.3 Testing the knee mechanism at slope movement

The knee mechanism was tested at slope climbing movement at 9 degrees slope inclination [26]. The reference trajectory was used to check the movement of the knee joint. PID Controller was utilized to explore the performance of the knee mechanism. Two time bases were selected to test the mechanism performance and the its dynamics. The output from the system was tested at 1 s and 0.1 s to get the performance at different times. As overall, the results at 1 s shows that the output track the desired input. However, at 0.1 s indicates a delay between the desired and the measured at the time interval. It is obvious that the generated torque and power that are required to move the system are 8 N.m and 1500 W respectively (Figure 13).

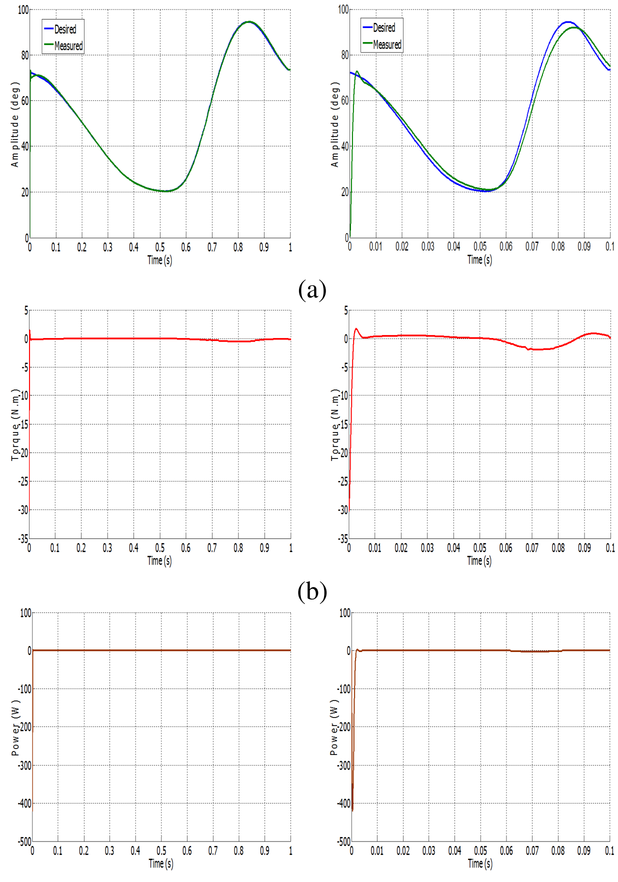
3.4 Testing the knee mechanism at stair ascent

The stair ascent movement of the knee mechanism was assessed by using a reference input [9]. The knee mechanism was tested at 1s and 0.1s with a desired input profile. Figure 14 shows that both time intervals successfully could track the desired movement trajectory. The torque and power generated from the actuator to the knee during the strides were recorded.



(a) (b) (c)

Figure 13. Knee angle, torque and power delivered from the actuator at slope climbing movement, 1 s and 0.1s, (a), (b), and (c).

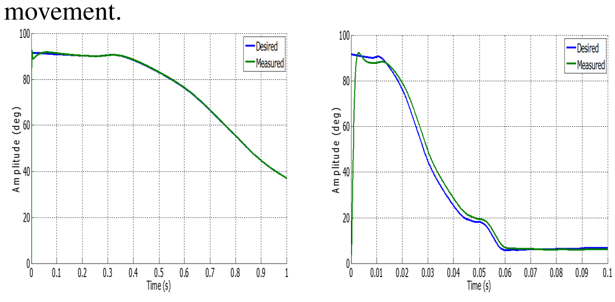


(a) (b) (c)

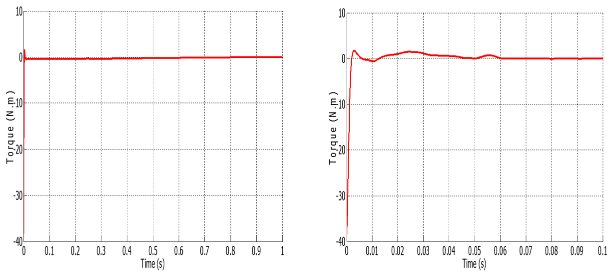
Figure 14. Knee angle, torque and power delivered from the actuator at stair ascent, 1 s and 0.1 s, (a), (b), and (c).

3.5 Testing the knee mechanism at sit- to- stand

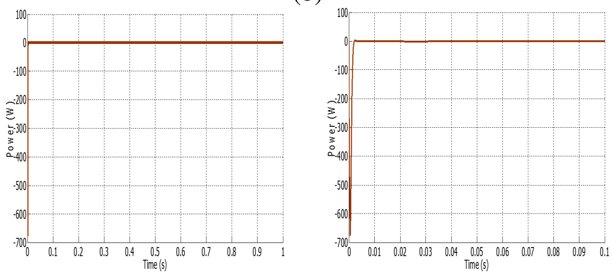
The performance of the knee mechanism at siting phase was tested according to a desired input [27] with time intervals of 1 s and 0.1 s the behavior of the system can be studied. The behavior of the knee mechanism during sit-to-stand was analyzed using PID controller. Figure 15 shows that the mechanism could respond to the desired inputs and both 1 s and 0.1 s. In addition, the torque and power were measured at the knee joint during the movement.



(a)



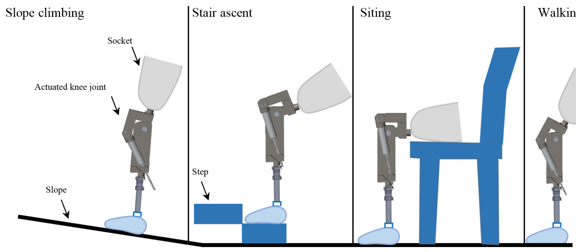
(b)



(c)

Figure 15. Knee angle, torque and power delivered from the actuator at sit- to- stand, 1 s and 0.1 s, (a), (b), and (c).

As the results of the actuated knee mechanism at various phases shown in Figure 12. It can be noticed, the flexion of the knee mechanism at 30°, 60°, and 90° showed zero error during the selected step time (0.2 s). However, the performance of the mechanism at 90° shows less settling time compared to the 30° and 60° phases. Therefore, it can be expected that the behavior of the knee mechanism at such sit- to- stand movement shall provide rapid response. The performance of the knee mechanism at 120° indicates a steady state error over the step time. The error is about 4.5° which is quite acceptable as the activity at 120° is seldom performed compared to the normal activities such as walking or sitting. Overall, the actuated knee mechanism showed an appropriate results at the four phases tested. The second part of the results present the capability of the proposed knee mechanism to track a desired or pre-defined input pattern at different time intervals. The behavior of the mechanism presented less error while tracking the desired input at 1 s and 0.1 s intervals respectively. On the other hand, at time intervals less than 0.1 the output results at different movements show less deviation from the desired. As the actuation system could not accommodate with fast movements according to the results. As the goal of the current study is to mimic the different movements of the knee joint by following the physical simulation goal achievement, the knee mechanism can be developed and components such the actuation system can be selected. Figure 16, shows the proposed knee mechanism of the actuated knee assembled with the other lower limb prosthetic components.



(a) (b) (c) (d)

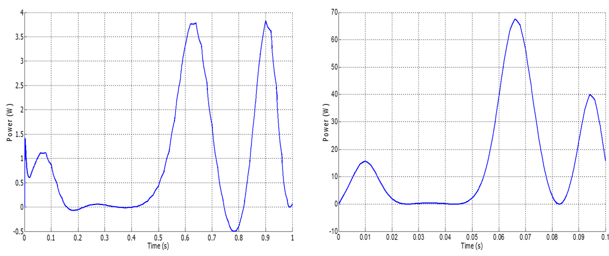
Figure 16. Complete structure of the knee mechanism at different activities, slope climbing, stair ascent sit- to- stand, and walking, (a), (b), (c), and (d).

The control of the actuated knee mechanism is necessary after considering all parameters and dynamics of the knee mechanism. In order to adjust the control parameters of the knee mechanism for walking, sit- to- stand, slope climbing, and stair ascent the parameters of the knee mechanism in terms of the spring stiffness and damping coefficient were adjusted as listed in Table 2. The PID control scheme could mimic the reference trajectory at each movement and could generate the sufficient force to the knee joint.

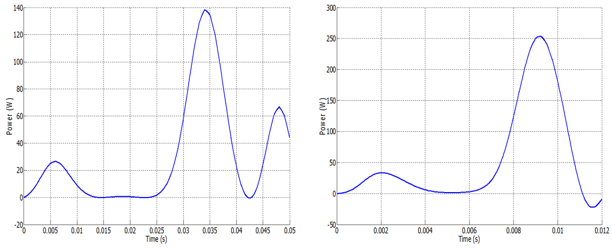
Table 2. Estimated parameters of the knee joint mechanism at average time interval of 1 s.

Activity	**Spring stiffness (N.m/deg)**	**Damping coefficient N.m/(deg/sec))**
Normal walking	4	0.003
Slope climbing	0.5	0.03
Stair ascent	0.8	0.1
Sit- to- stand	1.5	0.5

The actuation system is responsible to produce force at both direction (knee flexion and extension) in order to move the moment of arm of the knee. In the current investigation, the power variation provided from the actuation system was recorded at different time intervals (Figure 17). For the development it is required to estimate the average power required from the motor in order to move the system.



(a) (b)



(c) (d)

Figure 17. Estimated power required from the actuation system at 1s, 0.1s, 0.05 s, and 0.0125 s, (a), (b), (c), and (d).

Choosing the components of the proposed knee mechanism that fulfil the requirements of the adopted movements have to be concluded for physical implementation of the actuated knee mechanism. As the first prototype showed some lack in terms of time delay during the movement of the knee mechanism. Therefore, the actuation system was updated to adapt with the different knee movements. For instance it is expected that during the stance phase of normal walking the stiffness of the knee joint is high to maintain the knee in extension throughout the stance phase. Therefore, the actuation system controller must interact with the knee mechanism to vary its stiffness and damping at such situation. As the motor-spindle drive acts as an actuation system, the value of the spring stiffness during the design could be practically transformed to series spring component to improve the performance of the actuation system [15]. At the development stage and physical implementation of the actuated knee mechanism, there are varieties of choosing the actuation system that could actuate the knee mechanism. The simulation showed the proposed knee mechanism requires an average power delivered that from the actuation system at various time intervals as shown in Table 3.

Table 3. Average output power rates from the actuation system at various time cycle.

Time interval (s)	**Average motor power (W)**
1	1
0.1	12
0.05	24
0.0125	60

Based on Mechatronics approach, it is recommended to choose the sensory system in the same time with other parts of the system (actuation, mechanical, and control scheme) [28,29]. Thus, the sensory system should be selected in such a way to be compatible with the other components and should be integrated with other parts to come out with an integrated system [24]. The benefit of choosing the components of the system in the same time, may minimize the time of development and decrease some problems inside mechanical system such as friction [15]. The Mechatronics approach shall improve the overall performance of the system and make it more compact. There are various options of sensory system that could be used to measure the knee joint angle. Such sensors can be used in different locations either below the foot or integrated inside the socket of the amputee. The characteristics obtained from the current study concluded the main parameters requested to develop such actuated knee mechanism that could perform various movements whisch are listed in Table 3.

Table 4. Estimated parameters of the proposed actuated knee mechanism.

Parameter	**Value**
Peak actuator force	3500 N
Range of actuator force	-3000- 3500 N
Range of actuator motion	0- 0.1 m
Range of actuator velocity	0-14 m/s
Range of actuated knee mechanism	0- 120°
Average motor speed	8000 rpm
Average rated power	≈ 50 W
Estimated weight of the actuator system	0.3 kg
Estimated weight of actuated knee mechanism (Aluminium Alloy 6061)	2.02 kg

4 Conclusion

In this study, by means of physical modelling tool, overall features of an actuated prosthetic knee mechanism were obtained and controlled for the purpose of developing an actuated knee mechanism. Physical modelling tool was used to build the actuated knee mechanism and adjustment of all parameters and dynamics of the mechanism dynamics were performed. The angle of the mechanism was measured at all gait phases during walking according to a reference input signal. The actuated knee mechanism could rotate at a range from 0°-120°. The mechanism was tested under different time intervals with respect to the desired input angle. In addition, four knee movements (walking, stair ascent, slope climbing, and sit- to- stand) were presented and tested under different time bases. By adjusting the stiffness and damping at each movement, a PID control algorithm was built to control the behavior of the mechanism. The PID controller showed that the actuated knee mechanism could track the desired pattern at various activities. The parameters of the actuation system in terms of stiffness and damping were tuned and adjusted for the purpose of physical implementation of the actuated knee mechanism. Also, overall features of the actuated knee mechanism conducted from the physical modelling were obtained and listed. As can be noticed from the current study that the goal was to obtain a clear idea about the actuated knee mechanism features before the physical implementation. The physical simulation showed a realistic behavior of the knee mechanism and showed the workspace of the knee mechanism in terms of knee angle at different movements. However the current mechanism could not match with the desired movements at time less than 0.1s. Therefore, the designer should consider that issue at the development stage. In addition, the current mechanical design may be modified to assist the mechanism to perform different movements at less time intervals. Further investigation can be obtained from the actual physical system and the experiments. The evaluation of the system with amputee subjects shall provide better information about the system in real situation. Furthermore, it will assist the prosthetist to provide worthy feedback to the designer to improve the limitations of the current system.

5 Acknowledgement

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CHAPTER 4 : DISCUSSION

4.1 Introduction

The aim of the thesis was to show the possibility of using an alternative sensory system that is able to interact with the residual limb of the transfemoral amputee to better control the performance of the knee prosthesis in terms of the intention and transition between phases. This sensing system is supposed to be an alternative solution to electromyography (EMG) and electroencephalography (EEG) techniques that are currently used in some knee prostheses. Moreover, design, testing, and investigating of a knee prosthesis mechanism that could function within the normal range of about 120°and could replicate basic daily activities such as walking, sit-to-stand, stair ascent/descent, and slope climbing was sought. In order to address the aims of the thesis, the existing knee prostheses for transfemoral amputees was analyzed.

4.2 Outcome of the research questions

4.2.1 Sensory system

It is found that the existing sensory system in knee prosthesis for transfemoral amputees can be updated to better control its overall performance. Most studies of the knee prostheses used pure mechanical sensors to measure kinematics and kinetics parameters of the knee, for example, angle, force, and torque at the knee joint. Such sensors are fed to the control unit that consequently controls the movement and damping of the knee prosthesis during the movement. Beside the mechanical sensors, there are attempts to enhance the control of the powered knee prosthesis using EMG and EEG that are based on detecting the signals from muscle and brain activities, respectively. Previous types of knee prostheses (active/powered) are equipped with below foot on/off switches to detect the transition between states during the movement. The control unit of the knee prosthesis adjusts the amount of damping/torque according to the signals received from both mechanical sensors and the on/off switches.

Few critical points and questions are noted here:

a) Active/ powered knee prostheses contain on/off switches that are attached below the prosthetic foot to detect the transitions between phases: The user wears the knee prosthesis and a fixed foot prosthesis that contains on /off switches. The question that can be asked, is it necessary to find a method to transfer the on/off switches from the location below the foot to another location near the knee prosthesis?
b) The active/motorized knee prosthesis requires a sensing element to be in a direct contact with the human body (residual limb) to provide information about the amputee's intention (transition between phases). To what extent could the alternative sensing element that was adopted provide information about the intention or the transition between phases?
c) To what extent can the sensing element be used instead of the mechanical sensors that are used to measure force and torque of the knee?
d) Does the configuration of the sensing element affect the measurement of the output signal?
e) In terms of comfort and satisfaction, to what extent does the placement of the sensing element inside the socket's wall affect the satisfaction and comfort of the user?
f) To what extent could the sensing element be used as a pressure sensing element?

By addressing the above questions (a-f) and questions related to the sensory system that were adopted in the current thesis, the concept of using an embedded sensor that is placed inside the socket's wall was applied. Such location assists the sensing element to be directly in contact with the residual limb, or in direct contact with motor control system of the human body. It could also enhance the practical application of the system especially during donning and doffing, reducing the requirement of electrode placement and preparation such as in EMG and EEG applications. The sensing element is assumed to provide direct information about the transition during the movement of the knee prosthesis. There are existing studies that involve researches in the same field of the knee prosthesis that use mechanical sensors, EMG, EEG, and on/off switches techniques to control the knee prosthesis. Varol et al., (2009), developed a powered leg (knee and foot prosthesis), the controller of the powered leg was built, and the feedback was achieved using load cell and angle sensors that measure the force and angle of the knee respectively. The supervisory control system was developed and an embedded system was built. In addition, the phase transition was identified using on/off switches that are placed below the prosthetic foot. Although, there are attempts to enhance the performance of the knee prosthesis, still the mechanical sensors are used to measure the torque and position of the knee without using a reliable sensing element that could directly interact with the residual limb or amputee.

Another attempt by the researchers were to use EMG to improve the control the knee prosthesis. Dawley et al., (2013), used EMG approach to control the knee prosthesis, the knee prosthesis was tested with a unilateral transfemoral amputee. Although this study tested the performance of the prosthesis at level walking, the processing of EMG signals that generated from the muscles required an amplification and special calibration for the antagonist muscles.

Paper 1 discusses the concept of using in-socket sensor (sensor placed inside the socket's wall). However, the purpose was to improve the current sensory system of the existing knee prostheses. The alternative sensing elements could be involved in the field of the knee prosthesis to overcome the limitations of EMG and EEG techniques. The results obtained from the current study that are related to the in-socket sensors (piezoelectric bimorph as a sensing element) were promising, because, they showed the capability of piezoelectric bimorph to interact with the residual limb and successfully identify the transitions between phases during stride at various activities.

Paper 3 presents the usage of both FSR and piezoelectric bimorph as in-socket sensors. The obtained results are a promising step that respond to question b, it is considered that the transitions that were identified are occurred based on the change in the remaining muscles of the residual limbs. Afterwards, those signals from the change of muscles can be processed using some kind of classifier or pattern recognition algorithm. The classifier can be used to predict the intended action and feed it into the actuation system of the knee prosthesis.

Paper 3 refers also to the question a, based on the results obtained from the piezoelectric bimorph as in-socket sensor, and this preliminary investigation of using FSR sensor can be used instead of the below on/off switches. The FSR showed the possibility to be used as triggering sensor that could identify the transition between states. It is promising to the user to use knee prosthesis along with any type of foot prosthesis, especially if the lower prostheses (knee and foot prostheses) are consisted of two fixed modules which are permanently connected to each other such as Vanderbilt leg (Sup et al., 2011) . It is believed that it is easier and comfortable to the user to choose the type of foot prosthesis rather than being imposed to use a fixed type.

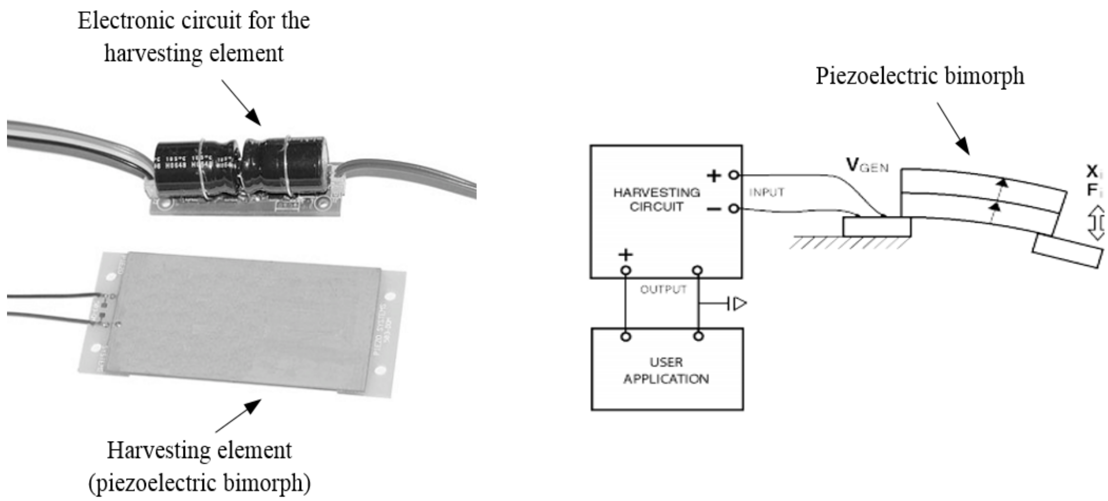
To discuss question c, the results acquired from the in-socket sensor cannot provide a comprehensive idea about the potential of using piezoelectric bimorph instead of the mechanical sensors. However, during the experiments that were conducted to compare FSR and piezoelectric bimorph, the piezoelectric bimorph showed an appropriate agreement with reference to the knee angle during the movements. But, further studies are recommended to be conducted using the piezoelectric bimorph to prove the possibility of using it as an alternative solution to the current mechanical sensors.

Further to question d and e, the configuration of the piezoelectric bimorph was selected based on the manufacturer dimensions, (Piezo systems, Inc, MA, USA). The characteristics that were obtained from the piezoelectric bimorph as a sensing and an actuation elements were based on those dimensions. However, if the shape and dimension of the bimorph are different from those that are used in the current study, it may lead to different static and dynamic characteristics, for example the trapezoidal and circular shapes provide different deflection and force values. Thus, it is recommended that the researchers to conduct a full calibration for the smart element that is going to be used in the future studies. Also, referring to question v, at the stage of conducting the experiments and trials, the amputee subject did not complain from discomfort or dissatisfaction due to the presence of the in-socket sensors (FSR and piezoelectric bimorph). It is considered that there is another positive indication of the feasibility of using the piezoelectric bimorph as an in socket sensing element.

Paper 1 discusses the full characteristics that were conducted with the piezoelectric bimorph beam as a sensing element which refers to question f. The results showed the capability of the bimorph to be used as an in-socket sensor. The case study that was conducted using piezoelectric bimorph showed that the piezoelectric bimorph could measure a maximum interface pressure of about 27 kPa that occurred at the anterior proximal site. The piezoelectric bimorph sensor was compared to the current available sensors, Flexforce and FBG in terms of rage of force and linearity. The piezoelectric bimorph showed similarity to the Flexforce sensor in terms of the static operating range, however the bimorph presented a more suitable dynamic measuring range compared to both the Flexforce (FSR) and FBG sensors. Therefore, it can be concluded that piezoelectric bimorph can be utilized as an in-socket to observe the pressure mapping between the socket and the stump.

It can be concluded that, a piezoelectric bimorph is considered a self-sensing element that is categorized under the sensory aspect of mechanoreceptors. In mechanoreceptors, signals come from the user to provide information about the user intention, and the signals are used to establish a robust controller for the prosthesis. Another advantage of the smart piezoelectric beams is their capabilities to function as a power harvesting device. The power harvesting is beneficial to produce some amount of power that can be stored to operate on/off switches or electronic circuits (Ottman et al., 2002). Therefore, the current study provides an overview about the usage of the piezoelectric bimorph as a power harvesting device. A question can be asked further on that point, how can the piezoelectric bimorph be used as a harvesting device and involved in the field of the knee prosthesis?

Paper 1 and paper 2 present the full characteristics of a piezoelectric bimorph as both a sensing and actuating elements were conducted. The piezoelectric bimorph as an actuator was tested with square and sine-wave forms, and the piezoelectric bimorph beam could bend in z- direction in a certain range of motion. Moreover, the experimental frequency response along the length of the piezoelectric bimorph is an important factor when the bimorph is utilized as a power harvesting device. The frequency response is set at the operating frequencies of the device, to avoid the resonance frequencies. Based on the information obtained from the piezoelectric bimorph characteristics presented in the current study, the circuit and configuration of how the piezoelectric bimorph element can be used as a harvesting device is shown in Figure 4.1.



(a) (b)

Figure 4.1: Piezoelectric bimorph harvesting kit, (a) Harvesting piezoelectric element circuit (Piezo Systems, Inc., MA, USA), (b) Harvesting electronic circuit

The harvesting circuit uses a piezoelectric bimorph can save about 5.2 V at the capacitor. When the output voltage reduces to 3.5 V, the circuit restarts the charging process. The basic concept of the circuit based on the energy harvesting bender (piezoelectric bimorph) during the compression and extension, one layer of the bimorph is compressed while the other layer is stretched, resulting in power generation. It may be excited by intermittent pulses or continuously from low frequency to resonant frequency (where rated displacement is achieved at the lowest force level). The piezoelectric bimorph as a harvesting device can be involved in the field of knee prosthesis. It is suggested that the arrangement of using the piezoelectric as a harvesting element can be used in knee prosthesis application, as shown in Figure 4.2. The diagram shows the framework of how the piezoelectric bimorph can be used as a sensing element as well as a harvesting device. The piezoelectric bimorph as a sensing element is called in-socket sensor that establishes a framework of controlling the knee prosthesis.

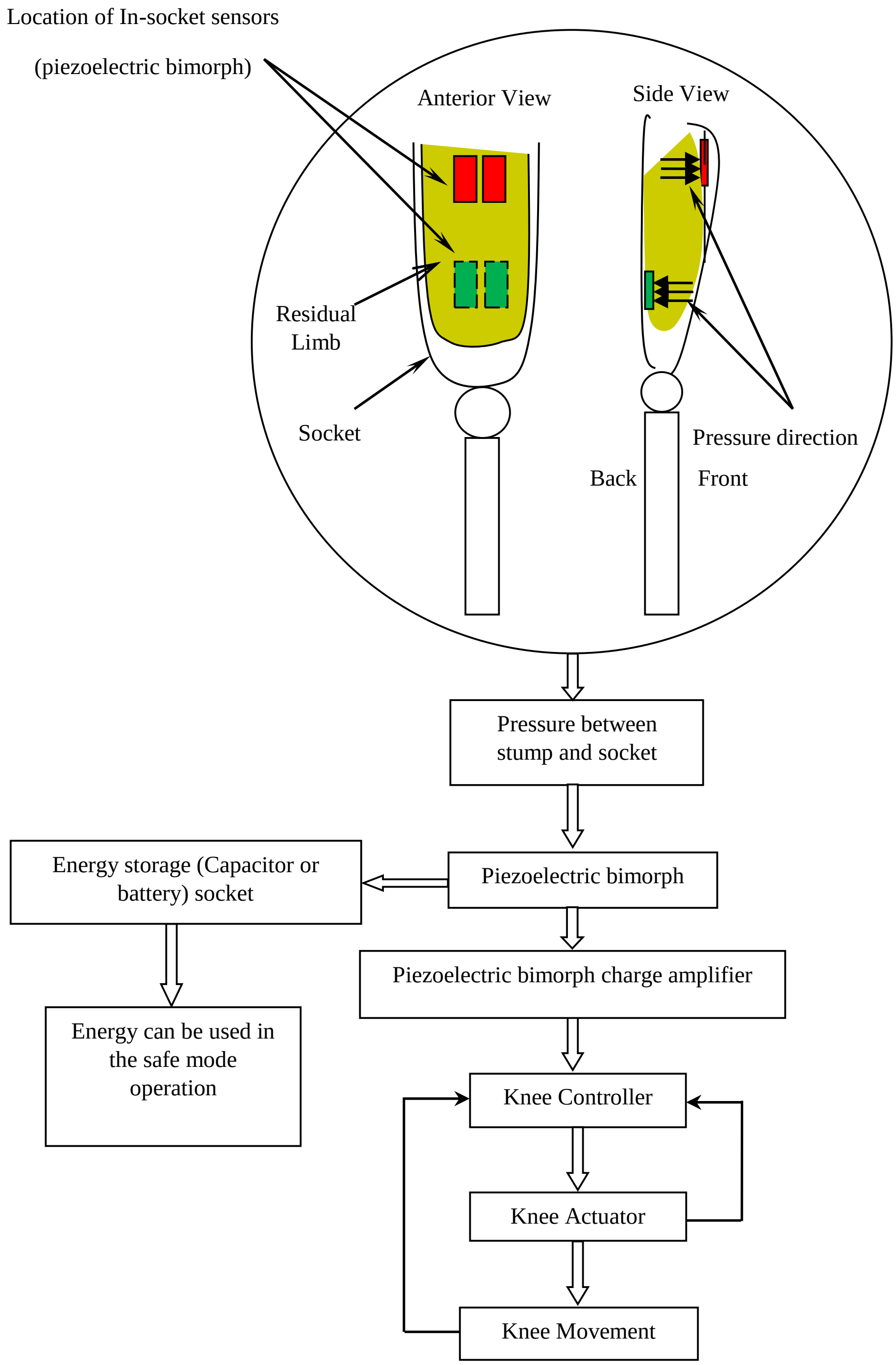


Figure 4.2: Diagram shows the possibility of using the piezoelectric bimorph as a power harvesting device besides controlling the knee prosthesis using piezoelectric in-socket sensor

4.2.2 Actuation system

Paper 4 presents the efficacy of the adopted actuation system of knee prosthesis to replicate the normal range of motion. The actuation systems of the current knee prosthesis were studied, various types of actuation systems are used in the knee prostheses. Herr et al., 2003, developed an active knee prosthesis that could replicate the walking pattern. Furthermore, pneumatic and hydraulic actuation systems were used in the early stage of developing the powered leg prosthesis by Sup et al., 2009. However, the prior types of the actuation system showed shortcomings in some daily activities that are requested by the user and still need to be achieved, for example, sitting to standing, and stair ascent/descent are not extensively studied. Another type of actuation system was used by Martinez et al., 2009, in which agonist-antagonist technique was used to provide movement to the knee prosthesis. Agonist-antagonist knee prosthesis was designed using an electric motor with series of springs. The knee prosthesis could mimic the normal walking of the amputee subject. However, no more trials were conducted to study the other daily activities. Goldfarb et al., 2011, developed a lower leg prosthesis using two electric motors for both foot and knee modules. The movement is delivered to the knee joint via a moment of arm using a spindle drive that is connected with the motor shaft. Experiments were conducted with an amputee subject to replicate walking, slope, and siting situation. The results showed that the leg prosthesis could assist the patient to do the previous activities.

Paper 4 also shows how the knee joint mechanism can be simulated and tested in order to be used in the knee prosthesis. The arrangement of the knee prosthesis was successfully simulated and tested for basic daily movements. The arrangement of the actuation system consists of an electric motor (EC) from Maxon®, combined with gearbox that is used as a knee actuator. A lead screw is connected to the motor/gearbox arrangement which delivers the moment of arm the required torque to move the knee joint.

A physical simulation tool using Simulink™, was adopted to simulate and test the suggested actuation system. The actuation system could operate the knee mechanism at basic daily movements such as walking, sit-to-stand, stair ascent/descent, slope climbing. Furthermore, the damping and stiffness of the actuation system were adjusted during each movement. The physical simulation was essential before the real development of the actuation system, as it provides a guideline to the people who are going to develop the lower prosthesis. Although, the actuation system was studied and simulated using the physical simulation tool, still the development and experimental study of the that proposed actuation system are needed for further analysis and investigation.

It is claimed that, the real development of the actuation system could not be achieved at the current stage of the ongoing research. However, an actuation system and mechanical structure were tested previously, but the system showed some limitations which are related to the low speed of the motor and the lead screw. Therefore, it is preferred to use the physical simulation tool to model, simulate, and test the actuation system before the development process. As a conclusion point of the knee actuation system, it can be referred to the human muscles that have a number of functionalities such as the generation, consumption and transmission of force, and energy storage. Thus, research towards new polymeric materials may be adopted for the field of the knee prosthesis. These materials could be used as artificial muscles that exhibit more muscle-like functionalities.

4.2.3 Mechanism and materials of the knee prosthesis

The third question is referring to the efficacy of the mechanism of the knee prosthesis to mimic the basic daily activities. The knee joint mechanism was designed, simulated, and tested at different daily activities. The knee mechanism is composed of the actuation system (linear actuation system) and the mechanical structure. As most activities occurred within the range from 0°-120° as reported by Kingston, 2001. The mechanism was designed using the CAD drawing software, Solidworks™. The CAD drawing was imported to the Simulink™ environment and the dynamics of the knee mechanism for all joints were determined and adjusted. The mechanism basically was designed based on the crank slider mechanism which was adopted by Varol et al., (2009) and Sup et al., (2007). The proposed knee mechanism could be made to provide a range of knee angle from 0°-120°. Also, a PID control algorithm was established to adjust the knee mechanism that is capable of delivering the required torque at basic daily activities (refer to paper 4). The different types of basic daily activities named, walking, sit-to-stand, stair ascent/descent, and slope climbing were tested within time intervals of 1 s, 0.5 s, and 0.1 s. The knee mechanism successfully replicated these movements at time intervals of 1 s and 0.5 s, respectively. However, the mechanical structure requires modifications to replicate the daily activities at a time interval of 0.1 s, as the values of the mass and inertia of the mechanical structure need to be revised to meet the requirements of a minimum time intervals.

In conclusion, the knee prosthesis mechanism was simulated and tested. The material of the knee structure was selected, Aluminum Alloy 6061as it is light in weight and less expensive compared to other alloys. Moreover, the mechanical structure should be fabricated with structural materials of an appropriate strength and durability as well as some functionality, also materials should give the prosthesis a lifelike natural body appearance (cosmesis). Furthermore, searching for new materials such as carbon fibre to be used in the prosthesis fabrication is recommended to improve the prosthesis functionality. Carbon fibre composite materials have the advantages of composite materials that include extremely strong and light weight, and are less-expensive than alternatives such as steel, aluminium, titanium and magnesium, and have greater resistance to corrosion, greater flexibility, impact resistance and vibration damping. They are also extremely resilient and show superior performance in a wide range of temperatures. The advantages of such new materials are savings in a patient's energy expenditure and an improved and more comfortable fit. Clinical studies have confirmed many aspects of these improvements. The following section shows an overall framework of controlling the actuated knee mechanism for transfemoral amputees using in- socket sensors.

Also, in order to develop an advanced knee prosthesis system, Figure 4.3 illustrates the general steps involved in system design and development of the knee prosthesis from initial stages to final validation and approval. It should be noted that all the stages of conceptual design, testing and validation of that, as well as detailed design and prototyping involving a number of iterations, before the final product satisfies the specifications and the requirements.

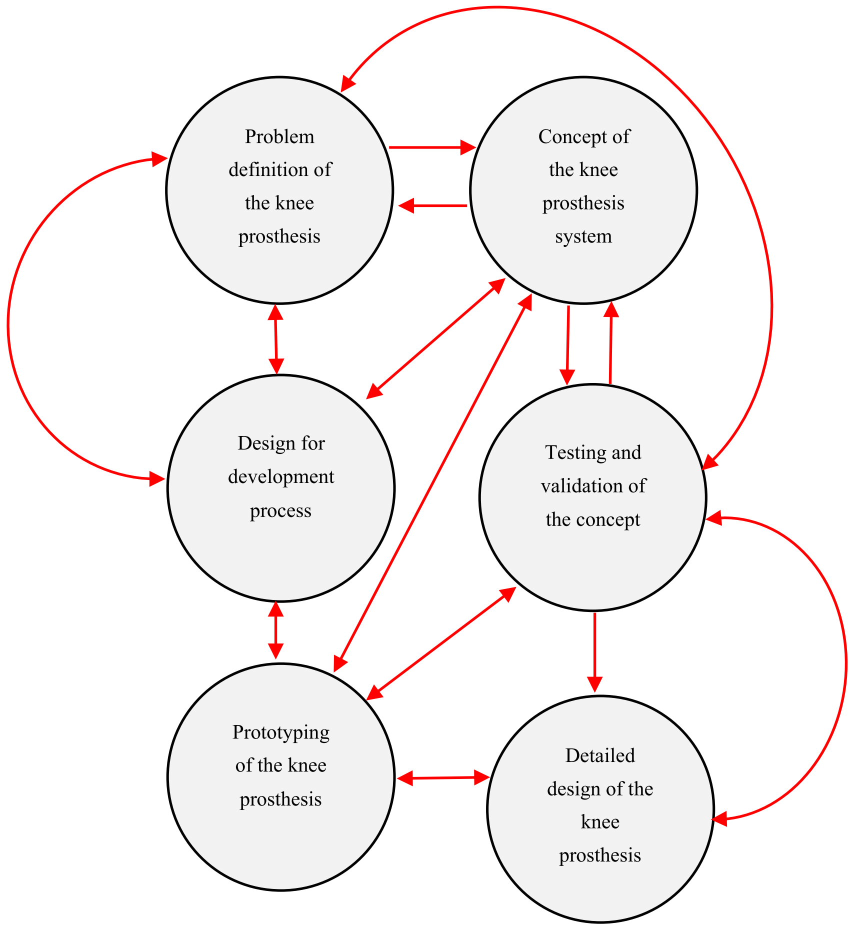


Figure 4.3: Suggested development stages of the future knee prosthesis

4.2.4 Framework of controlling the knee prosthesis using in-socket sensor

This section presents the possibility of controlling the knee prosthesis using the in-socket sensors (Figure 4.4). The prosthetic limb can be integrated with the in-socket sensory system, the sensing element is a force sensing resistor or a piezoelectric sensor. The sensory system can detect the gait phase comprises heel strike, foot flat, toe-off, stair ascend, and sit to stand movements. The prosthetic limb integrated with the sensory system, the sensing element is mounted on an inner surface of the prosthetic socket.

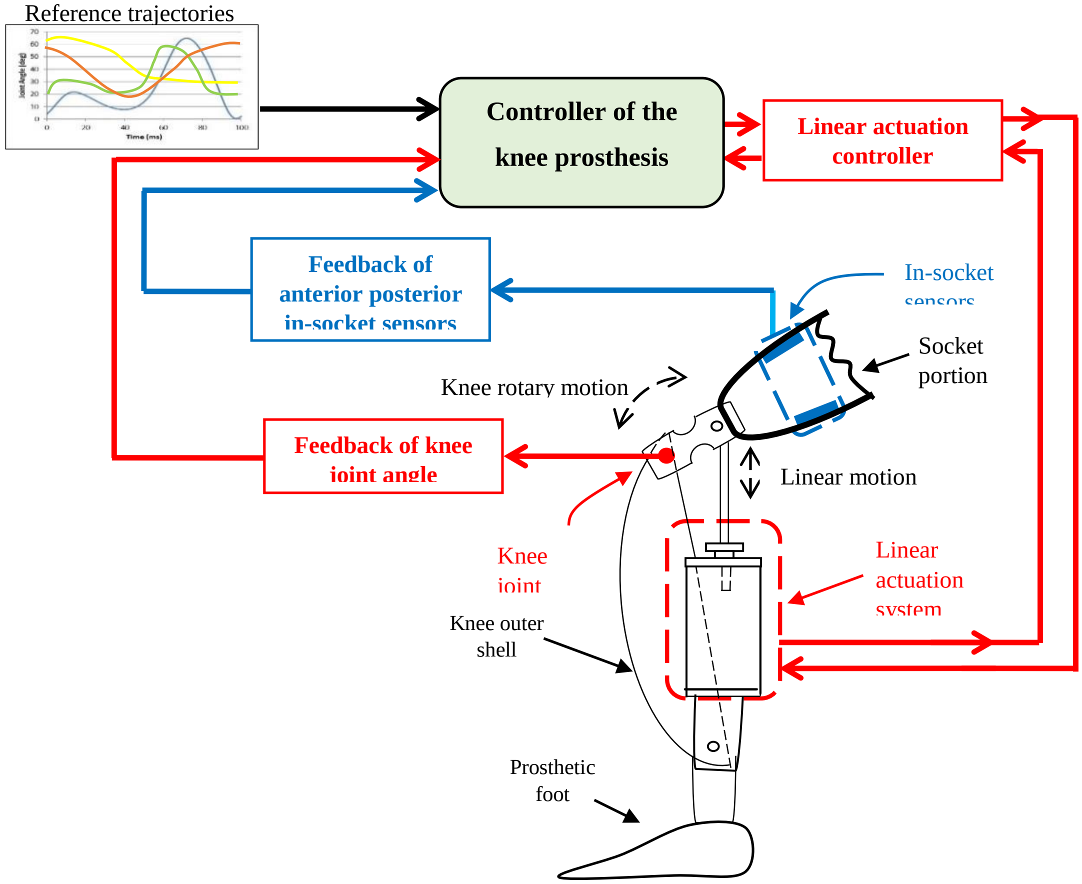


Figure 4.4: Detailed description of the controlling the prosthetic limb via in-socket sensors

The prosthetic limb integrated with the sensory system, the sensing elements are mounted at an anterior rectus femoris and posterior biceps femoris regions of the prosthetic socket. The method of producing the prosthetic limb movement by using the prosthetic limb integrated with the sensory system can be performed according to the following steps. Firstly, transmitting information from the sensing element to the controller. Secondly, estimating the gait phase using the controller and creating the input signal based on the information received from the plurality of sensing elements. Thirdly, sending the input signal from the controller to the actuator to produce the prosthetic limb movement.

The information of the lower limb can detected by the sensing element and then analyzed by the controller based on maximum value of amplitude voltage, minimum value of amplitude voltage. The prosthetic socket can be made of various materials and in different sizes, but preferably custom-made for each amputee according to the shape and condition of the residual amputee stump and the amputee's mobility grade.

A lever arm connects the prosthetic socket to an actuator for actuating a movement of the prosthetic limb. The actuator is an electric motor coupled with a spindle drive and revolute joint on a knee chassis. The spindle is connected to a spindle nut that moves along the spindle to produce a linear motion of the prosthetic limb as shown in Figure 4.4. The prosthetic limb further includes a fail-safe feature in which the prosthetic limb will become a free moving passive limb or will fully extend in the case of a system failure or a loss of power supply. This will allow the amputee some level of mobility and stability to get to a safe place to assess a problem or to get help.

The prosthetic limb is integrated with the sensory system for detecting amputee’s intention movement in order to achieve a more responsive gait as well as improve abilities beyond normal walking such as stair climbing and hill descent. The sensory system is a communication mechanism connecting the sensing element, the controller, and the actuator. The sensing element is mounted on an inner surface of the prosthetic socket. Generally, placing the sensing element in these two regions (biceps and rectus femoris) should be sufficient in detecting the amputee’s intention for movement. Adding the sensing element onto those regions at both proximal and distal regions of the prosthetic socket may give a more accurate reading for estimation of a gait phase.

CHAPTER 5 : CONCLUSION

In the current thesis three aims were presented, each aim was achieved at a specific stage of the research. First stage, the smart piezoelectric bimorph was fully tested in terms of its functionality as a sensing element and also an actuation element. The piezoelectric bimorph was characterized as an in- sensor, and trials were conducted with an amputee subject. The bimorph showed the ability to measure the pressure variation during the stride with comparison of FSR and FBG. In addition, the bimorph static and dynamic parameters were worked out to be used as guidelines for the future development of the knee prosthesis. Similarly, the capability of piezoelectric bimorph as an actuator was tested by using it in a particular application of grasping small object. The deflection, time response, and dynamic range of the piezoelectric bimorph were experimentally investigated to show the validity of the piezoelectric bimorph at this particular application.

The second stage was to compare the piezoelectric bimorph as in–socket sensor with another available sensor force sensitive resistor (FSR). The comparison was conducted by utilizing the piezoelectric bimorph and the FSR to detect the movements of the knee prosthesis during walking, stair ascent, and sit-to-stand. Both sensing elements were embedded to the socket’s wall, and the subject was asked to perform different knee activities. The piezoelectric bimorph showed agreement with reference to the knee angle at most of the movements. However, the FSR was exhibited as a triggering sensor while the knee angle varied during the movements, therefore, it is recommended to use the FSR as a triggering sensor.

The third stage of the current research was to design, simulate, test, and control a new knee prosthesis mechanism that is actuated by a linear actuation system. As the literature indicated, some limitation of the actuation system in terms of generating required torque and power at movements such as stair ascent, stair climbing, and sit-to-stand as well as some limitation in the range of motion. Thus, the linear actuated knee mechanism was physically simulated, and its performance during different basic daily activities was tested. The new actuated mechanism was capable of replicating the movements while the knee mechanism was controlled using PID controller, the actuation system could generate the required torque at different knee flexion angles. The parameters of the actuation system and the estimated parameters of the knee prosthesis mechanism were listed for the purpose of future development of the mechanism. The physical simulation tool was an efficient method while the dynamic parameters in terms of damping coefficient, spring stiffness, and inertia were incorporated during the simulation. In addition, the physical simulation provides more flexibility to test the knee mechanism before the development stage.

It is believed that there are possibilities to improve the work presented here. Furthermore, several important directions for the future work can be identified. However, it is seen that the first step to further improve the current work, is to assess it based on the following three points:

a. The aim of the work:

The current work presents two aims which are, design, testing, investigation and evaluation of the sensory system, and mechanism of the knee prosthesis. First, with respect to the design of the sensing element which is the basic part of the sensory system was not fully achieved in the current thesis. However the piezoelectric bimorph was purchased from the manufacturer for the purpose of using it in the prosthesis application. In addition, the signal conditioning circuits and data acquisition system were established during the experiments. The piezoelectric bimorph is considered a smart sensor that can function without the need to external power supply. Second, the testing and investigation of the piezoelectric bimorph were carried out. A comprehensive calibration was performed using a standard calibration machine. Also, investigation of the sensory/actuation capability of the bimorph was achieved by using it in two different applications, to evaluate its performance. Moreover, the full investigation that was investigated will provide an appropriate guideline to the researchers to make use of such kind of smart elements in the field of lower and upper prosthesis. Finally, the clinical evaluation of the piezoelectric bimorph was performed by fixing it inside the socket's wall of a transfemoral amputee, while the amputee was instructed to replicate different daily activities. Each activity was repeated 5 times to check the repeatability and sensitivity of the sensory system in the real situation. It is believed that, more clinical trials may give further evaluation of the sensory system.

The second aim is related to the prosthesis mechanism. The design of the knee mechanism was successfully accomplished and simulated. The investigation of the mechanism was simulated using a physical simulation tool. This simulation tool is a new approach that is used nowadays in simulating different control system applications. The physical simulation could mimic the realistic behavior of a system with acceptable assumptions. However, still the real development of the knee mechanism may show a realistic investigation than the simulated one. At the current study, the knee joint mechanism could not be fabricated. However, the first prototype was developed at the early stage of the research. The prototype showed some limitations in some aspects such as speed of the knee rotation and mechanical design. So, it is recommended to use the physical simulation to check the validity of the knee mechanism prior to the development process. Although the physical simulation could successfully mimic the performance of knee mechanism during simulation, the knee mechanism has not been yet tested by a user. Testing the knee mechanism with transfemoral amputees has to be taken into consideration in the future research.

b. The significance of thesis contribution

The thesis presents some important aspects that may be useful in the field of prostheses development First, the possibility of using the smart piezoelectric bimorph as an alternative sensing element to EMG and EEG in the field of prostheses. The literature does not provide sufficient characteristics about the sensing element (piezoelectric bimorph) that was used in the current study. Thus, full characterization and evaluation of the piezoelectric bimorph that was conducted during the study is considered a crucial step in the field of prostheses. On the other hand, the physical simulation tool using Simulink™ is newly used in the field of designing and developing the prostheses especially in the analysis of the system's dynamics and building the control system. This physical simulation tool reduces the errors that may occur during the manufacturing process of the mechanism and it provides feedback to the designer for updating the dimensions of the mechanism according to the suggestions and recommendations received from the users and specialists.

c. Appropriate methodology and methods

In the current thesis, a non-traditional philosophy was adopted in order to achieve the aim of the thesis. It is believed that the biomechatronics philosophy provides a possibility to use a smart sensing element that was involved in the in-socket sensor and simulate and test the knee mechanism. Furthermore, the standard calibration machine that was used to perform the full calibration for the sensing element contains useful static and dynamic functions. Especially when performing the dynamic test for the sensing element, the machine could generate various types of functions such as sine and square waves with different amplitudes and frequencies. Those functions helped to get extensive characteristics of sensing that was not fully performed in the previous studies. On the other hand, the type of socket that was used during the experiments which was a quadrilateral double socket, was used by the user for several years. Also, it was informed by the user that his socket was comfortable and he was satisfied with it. However, different types of sockets are recommended to be tested for further clinical evaluation of the sensory system.

5.1 Directions for Future Work

5.1.1 Knee prosthesis design and development

Design and development of the knee prosthesis should start with a clear description of requirements, which will be translated into system specifications. For lower limb prosthesis, the aim is a system that mimics human locomotion, including all aspects of functionality and appearance. By considering the functionality, system requirements may be summarized in general terms as follows. The system should provide locomotion like normal, interact with the user, be comfortable, and adapt with different terrains and environments,

Research is needed to clearly understand normal human locomotion and how the biological system adapts itself to various terrains and environments. Such research into normal and pathological human locomotion has greatly helped the design and development of new generations of the prosthesis. The testing and evaluation of these systems on amputees and the feedback from the various parties involved to have also helped in system development and improvement.

5.1.2 Useful points for the upcoming research

As a final conclusion of the current study, it can be noted that there are several possibilities to enhance the current work and various directions of further research that can be listed as follows:

i) Clinical trials with more subjects of different types of the socket will provide more investigation about the behavior of the in-socket sensor during different level of movements.

ii) The development, testing, and clinical evaluation of the adopted new design of the knee mechanism with a transfemoral amputees, may provide some important aspects that have to be considered during the early stage of the design.

iii) Development of knee prosthesis using new materials such as carbon fibre will lead to light weight and less-expensive knee system compared to other types of materials.

iv) The framework of controlling the actuated knee mechanism can be implemented and tested with different amputee subjects, various types of sockets, and different level of amputations.

v) Piezoelectric bimorph and FSR sensors that were used in the current study may be compared with different type of sensing technologies such as F-scan for more comprehensive investigation about the effect of using the variety of sensors.

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