DEVELOPMENT OF KNEE JOINT FOR A MOTOR-ACTUATED KNEE PROSTHESIS

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Abstract

A well-designed lower limb prosthesis is required in order to allow an amputee to return to healthy locomotion. A complete set of the prosthesis for transfemoral amputee includes a socket which is custom fit to the residual limb, a prosthetic knee unit, a shank connecting the knee unit to a prosthetic ankle, and a prosthetic foot. The purpose of this project is to improve the initial design of a microcontroller-based transfemoral prosthesis which was unable to reach the designated rotational speed intended for a functional use, and had low aesthetic value. The improvement focuses on designing a new mechanism on the knee unit by using a pre-existed motor, a socket, and the prosthetic foot available from the previous design. This includes mimicking the normal limb function and to enhance the range of motion of this knee prosthesis. For aesthetic reason, a new cover of the shank will be designed. There are a few criteria that need to be fulfilled at the end of this project such as the new proposed knee unit should be able to bend up to 120^{O} and has enough strength to withstand the amputee's body weight. A computer aided design (CAD) software known as SolidWorks 2013 was used to design every single component of the knee unit and the shank cover. Then, each component will be assembled and the mechanism of the knee unit will be simulated in order to visualize the expected motion or movement of this new knee prosthesis. The materials used was selected by considering the cost and the properties of that material. Finite element analysis (FEA) was performed and the results were evaluated to determine if the proposed prosthesis can withstand certain loads. In the end, all results were discussed and evaluated to check if the design has fulfilled the goals of this project.

Abstrak

Kaki palsu diperlukan oleh individu tanpa kaki untuk membolehkan mereka bergerak secara normal. Satu set kaki palsu yang sempurna harus mempunyai soket, unit lutut palsu, betis palsu, dan tapak kaki palsu. Projek ini dilaksanakan untuk menambah baik rekabentuk pertama kaki palsu yang tidak mencapai kelajuan yang diperlukan untuk melipat kaki palsu tersebut. Ia juga mempunyai nilai estetik yang rendah. Penambahbaikan difokuskan kepada merekabentuk satu mekanisma baru dengan menggunakan motor dan soket yang sedia ada. Rekacipta tersebut haruslah meniru fungsi kaki manusia yang normal serta dapat meningkatkan sudut semasa melipat kebelakang unit lutut palsu tersebut. Nilai estetik unit lutut palsu tersebut harus ditingkatkan dengan merekabentuk satu kerangka baru. Reka bentuk kaki palsu tersebut haruslah mempunyai keupayaan untuk dilipatkan sehingga mencecah 120 darjah and mempunyai kekuatan untuk menampung berat badan mana-mana individu tanpa kaki. Perisian rekabentuk berbantukan komputer (CAD) yang dikenali sebagai SolidWorks 2013 telah digunakan untuk merekabentuk setiap bahagian lutut palsu tersebut. Mekanisma pergerakan unit lutut palsu tersebut telah disimulasikan untuk melihat bagaimana ia bergerak. Bahan yang digunakan untuk membina unit lutut palsu tersebut akan dipilih dengan mengambilkira kos dan ciri-ciri bahan tersebut. Analisa elemen terbatas (FEA) telah dilakukan ke atas unit lutut palsu tersebut dah keputusannya telah disemak untuk memastikan ianya mampu menampung beban badan mana-mana individu tanpa kaki. Pada akhir projek, kesemua keputusan telah dibincangkan untuk menyemak sama ada rekabentuk tersebut telah memenuhi matlamat projek ini.

Acknowledgement

I would like to take this opportunity to express my profound gratitude and deep regards to my supervisor, Dr. Nur Azah Binti Hamzaid for her exemplary guidance, monitoring, and constant encouragement throughout the course of this project. The blessing, help, and guidance given by her time to time shall carry me a long way in the journey of life on which I am about to embark. Without her supervision and friendly help this project would not have been possible.

I want to express my deep gratitude to my parents because of their understanding, encouragement, and personal guidance which provided a good support for this thesis project. The countless times they have spent to fire up my spirit during my hectic schedule will not be forgotten.

Lastly, to all family members, friends and coursemates, all your supports are gratefully acknowledged.

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List of Symbols and Abbreviations

Acrylonitrile butadiene styrene	**ABS**
Agonist-antagonist active knee prosthesis	**AAAKP**
Computed Aided Design	**CAD**
Finite Element Analysis	**FEA**
Lower limb amputation	**LLA**

Chapter 1. INTRODUCTION

1.1. Overview

1.1.1. The Knee Joint

The knee is where the upper and lower legs join together. It plays crucial roles in the body, and provides the bending motion that allows us to walk. It is the largest synovial joint in the body. It consists of two joints which known as tibiofemoral joint (an articulation between femur and tibia), and patellofemoral joint (an articulation between the patella and femur) as illustrated in Figure 1.1. The knee is one of the most mechanically vulnerable areas. It is due to the forces and leverage which act on the joint both in terms of chronic weight bearing problems, and also in terms of traumatic impacts (Kingston, 2001).

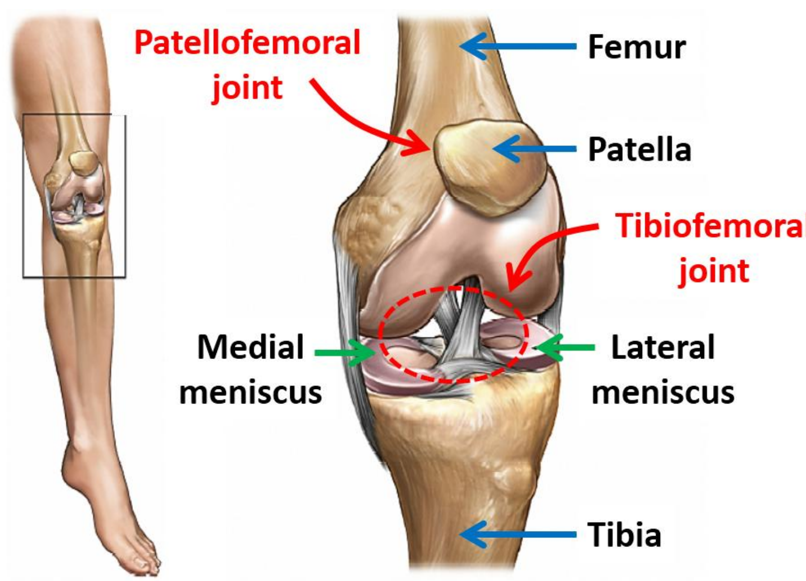


Figure 1.1: The knee joint (Adapted from NucleusMedical Art, Inc., 2012)

Two major movements of the knee are flexion and extension as illustrated in Figure 1.2. Meanwhile, Figure 1.3 illustrates on how the bicondylar mechanism of the transfemoral joint. This joint allows flexion and extension through the transverse axis with limited rotation through longitudinal axis (Kingston, 2001)

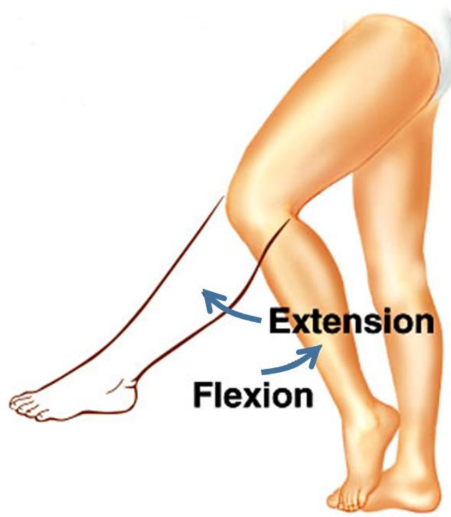


Figure 1.2: The flexion and extension of the knee joint (Adapted from Wikipedia, 2013)

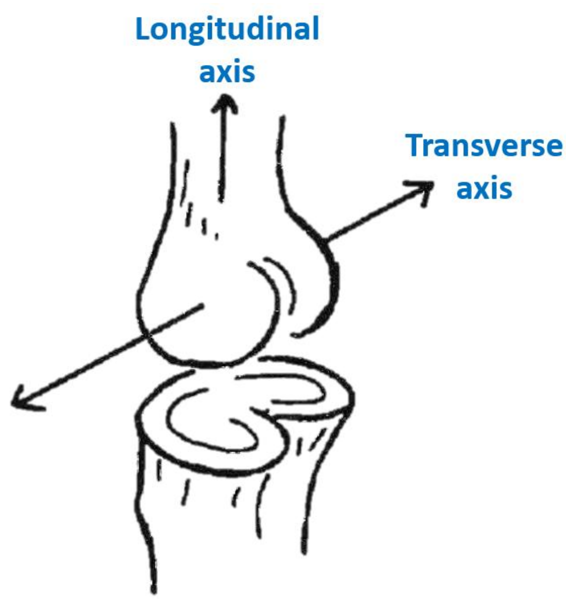


Figure 1.3: The axes of transtibial joint (Adapted from Kingston, 2001)

1.1.2. Lower Limb Amputation

An amputation can be defined as removal of the limb or outgrowth of the body (Dorland & Anderson, 2003). A decision to amputate someone is made when an injured limb cannot be recovered to function after disease or trauma. An amputation is also known as a life-saving operation when no other choices available. It must not be considered as a failed treatment. The amputation is one of the oldest surgical procedures that has been practiced by surgeons for decades. The development of prosthetic design and specific rehabilitation programmes for people with limb loss was intensified after World War II (Bowker & Pritham, 2004).

A lower limb amputation (LLA) can be categorized into a major or minor amputation. A major amputation is performed proximal to the ankle joint. Meanwhile, a minor amputation is one performed distal to the ankle joint. Transfemoral amputation, knee disarticulation, and transtibial amputations are three most common levels of amputation as illustrated in Figure 1.4. Most of the LLA were reported due to peripheral vascular disease. Nowadays, this incidence involves 12 to 44 per 100000 persons annually. A person with diabetes mellitus has a higher risk of amputation compared to a normal person (Ephraim & Dillingham, 2003). One of the main concerns with the level of amputation is the energy cost of the prosthetic limb gait. A higher level amputation requires more energy during ambulation compared to the ambulation of lower level amputation (Waters & Mulroy, 1999).

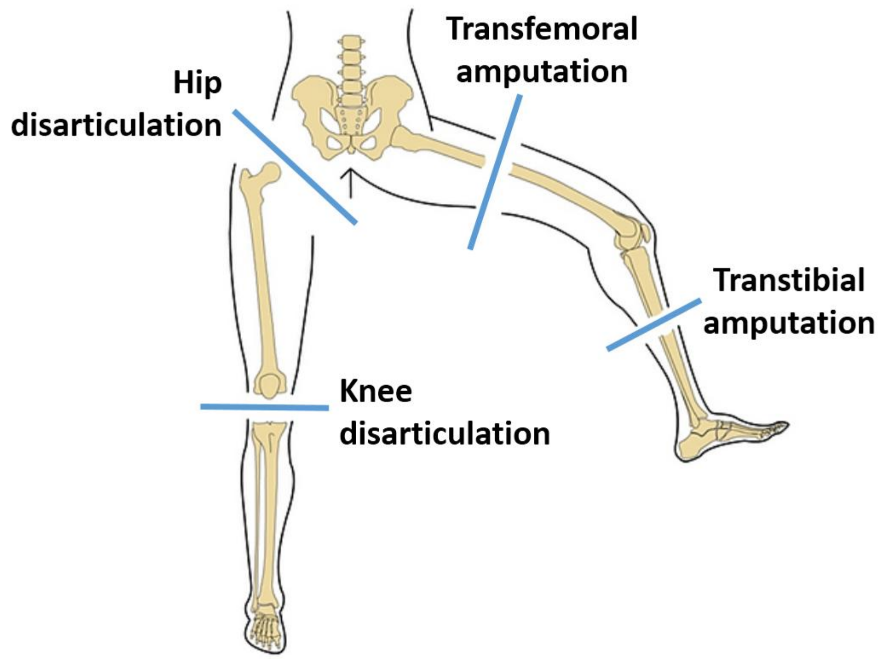


Figure 1.4: Lower extremity amputation levels (Reproduced from Burns, 2010)

1.1.3. Human Gait Cycle

Walking is defined as a method of locomotion involving the use of two legs in order to provide support and propulsion. A repetitive sequence of limb motion is required to move the body forward while simultaneously maintaining stance stability. Gait represents the manner of walking of any individual, rather than the actual walking process. The gait cycle is known as a period of time between two consecutive heel contacts of the same foot. Let say, if someone starts with the right foot contacting the ground, then the cycle ends when the right foot makes contact with the ground again. There are two phases in the cycle, which are stance phase and swing phase.

The stance phase is defined as part of the cycle when the foot is in contact with the ground, which comprises of 62 percent of the cycle. It begins with initial foot strike and it ends with toe-off. Meanwhile, the swing phase is known as an interval in which the foot is not contact with the ground. It comprises of 38 percent of the gait cycle. The swing phase is the phase when all portions of the foot are in forward motion. Figure 1.5 is the diagram to represent the gait cycle. The gait pathologies and walking speed affect the period of time to complete the gait cycle (Larsson et al., 1980).

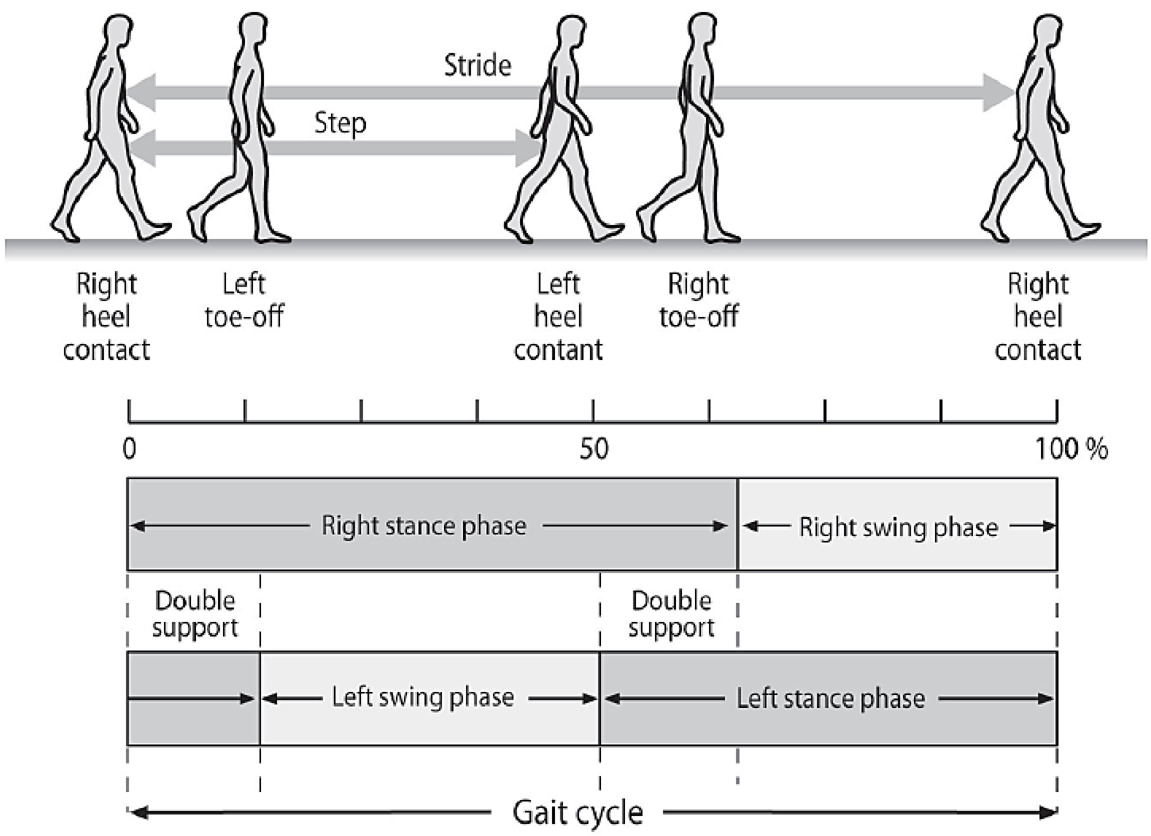


Figure 1.5: The normal gait cycle (Reproduced from Liikavainio, 2010)

For transfemoral amputees, gait patterns change in comparison with normal gait due to absence of muscles in the prosthetic limb. It leads the prosthetic limb to have uncontrolled flexion during stance phase. This can be avoided by substituting the prosthetic limb with a substitute stabilization device. At the same time, a prosthetic knee should be able to allow and control mobility between the thigh and lower leg during swing phase (Sapin & Goujon, 2008).

1.1.4. Types of Prosthetic Knees

The quality of life of an amputee can be improved by fitting his/her residual limb with an appropriate prosthetic leg. A common prosthetic leg for a transfemoral amputee comprises a custom fitted socket to the residual thigh, a knee joint, an ankle, a foot, and some connecting components as illustrated in Figure 1.6. A good prosthetic knee should mimic the behaviour of the biological knee.

Nowadays, many types of prosthetic knees are available in the market. They come from simple, purely mechanical devices up to sophisticated microprocessor controlled systems. Typically, a mechanical knee trades increased mobility for decreased stability (Lambrecht, 2008). This is where a prosthetist plays his role in helping the amputees to find a suitable balance for the individual amputee’s muscular coordination. The types of prosthetic knees are summarized in Table 1.1.

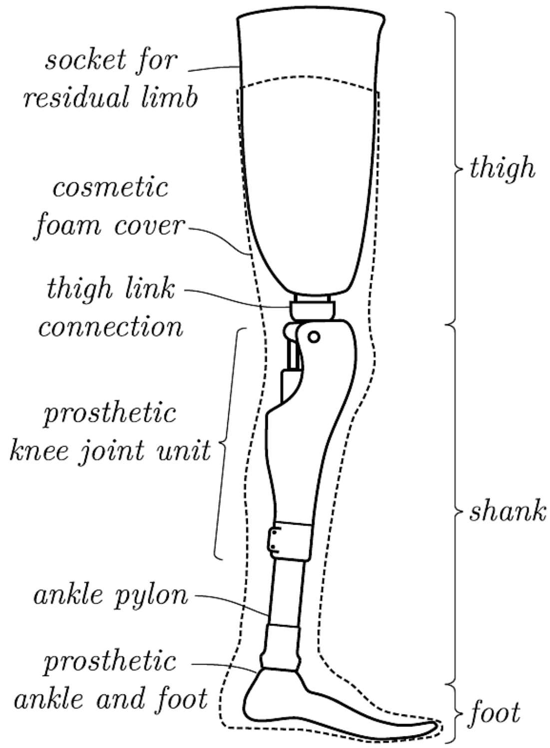


Figure 1.6: A complete prosthetic leg for transfemoral amputee (Reproduced from Lambrecht, 2008)

Table 1.1: Types of prothetic knees (Adapted from Lambrecht, 2008)

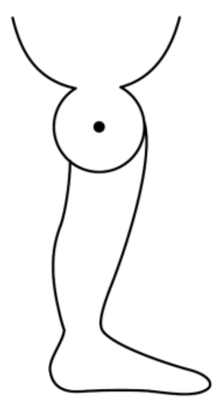
Types of prosthetic knees	**Descriptions**
Single axis knee 	• The most basic prosthetic knee. • Comprises a single pivot between the thigh socket and shank. • As long as the leg is straight at heel strike, the knee will not buckle during stance. • The amputee may need to exert more hip extension torque to keep the knee over center during mid to late stance.

Table 1.1, continued

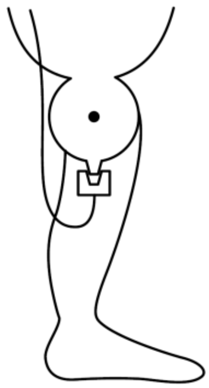
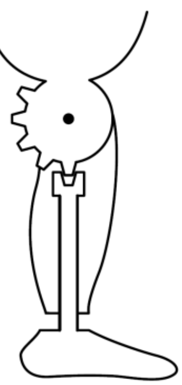
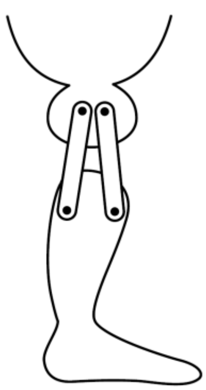
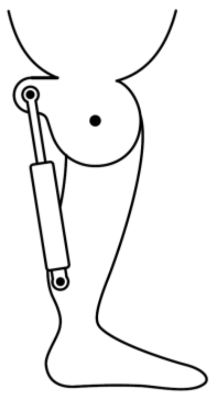
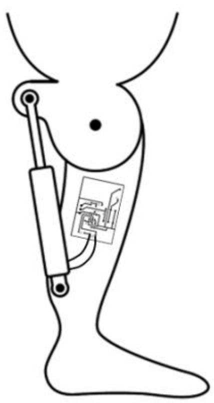
Types of prosthetic knees	Descriptions
Locking knees 	• Kept straight during walking. • Incorporates a cable or lever release to allow the knee to bend for sitting. • A locking knee is fitted when amputee lacks the hip control to stabilize a bending knee. • May cause inefficient still legged gait. • No chance of the knee buckling and causing the amputee to fall.
Stance control knees 	• Single axis knee with friction break (self-lock). • A load from the amputee's body weight makes the friction brake engages and preventing the knee from buckling. • The brake disengages when the load is removed, which allows the shank swings freely. • It does not allow knee flexion in terminal stance to initiate swing until the knee is fully unloaded.
Polycentric knees 	• Consists of four bar mechanism which move in the sagittal plane. • Near the straight position of the knee, the effective pivot lies at the posterior of the load path for safe early stance loading. • During terminal stance, it buckles easily to initiate the swing.

Table 1.1, continued

Types of prosthetic knees	Descriptions
Pneumatic/Hydraulic knees 	• Consists of a hydraulic cylinder in a single axis or polycentric knee design. • The cylinder contains valves which restrict the fluid flow between the two sides of the cylinder. • The adjustable valves allow the prosthetist or amputees to adjust the free swing flexion and extension for comfortable walking speed.
Microprocessor knees 	• Incorporate angle and load sensors with an on board microprocessor to adapt dynamically to change gait. • Flexion/extension is adapted by a servo controlled valve. • Some microprocessor control only during swing phase. Meanwhile, others controlling both stance and swing.

1.2. Scope of Work

A study of a microprocessor prosthetic leg is being carried out by a group of researchers from the Department of Biomedical Engineering, University of Malaya. The study aims to develop a bionic leg that detects user intention, to create a leg that can think by itself, and to develop knee with controlled action. The main objective of that study is to design a new microcontroller prosthetic knee with implanted piezoelectric materials at the socket as illustrated in Figure 1.7. That is the big picture of the study, while this thesis project focuses only in developing the mechanical structure of the prosthetic knee joint in terms of improving the performance of the initial design, and to increase the aesthetic value of the bionic leg. The materials of each component of the prosthetic knee joint will be selected carefully. The prosthetic knee joint was simulated to study its strength, and mobility.

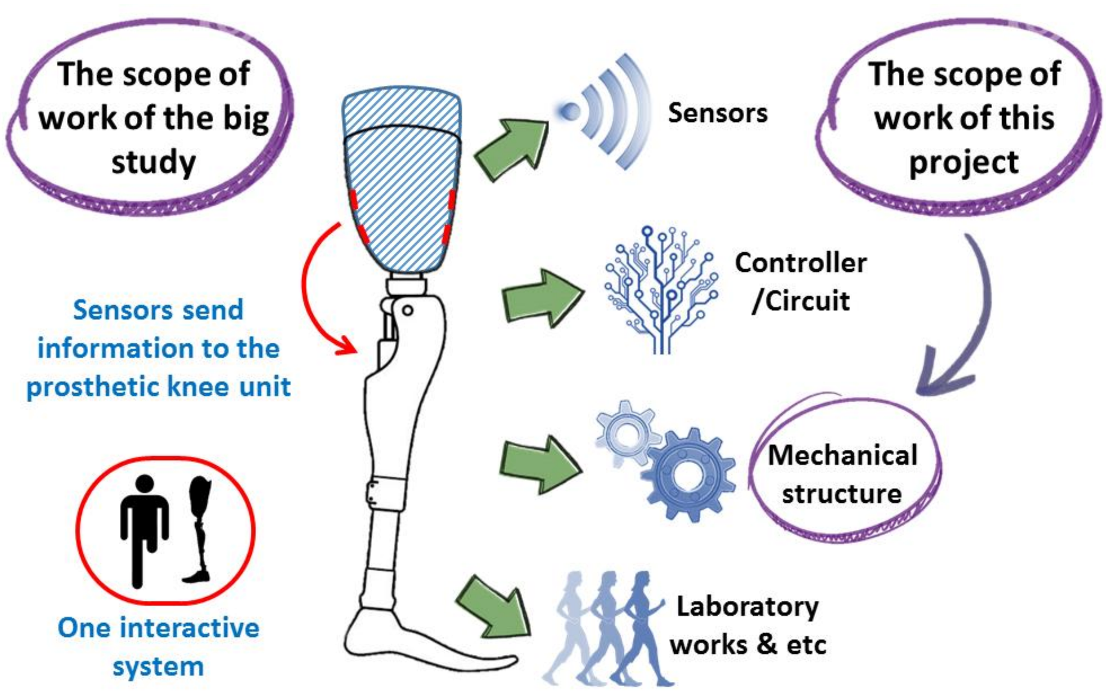


Figure 1.7: The schematic diagram of the scope of work for this project

1.3. Problem Statement

There are a few things of the initial design that needs to be improved. The actuated motor was purchased by focusing too much on torque of the actuated motor. The speed of the motor seemed to have been overlooked. The initial design flexes in a very slow motion. It takes approximately 40 seconds to bend the knee from 0 – 90^{O}. The prosthetic knee should be able to flex very quickly in order to allow the amputee to move fast. Furthermore, the range of motion up to 90^{O} does not mimic the range of motion of human biological knee. The length of the initial design was approximately 450 mm. It is bulky and unsuitable to be used for Malaysian amputees. The long cylindrical shell of the initial design does not look appealing. To allow the prosthetic knee to look nicer, a new cover of the prosthetic knee was proposed. Although the speed of the available actuated motor is very slow, the proposed design of the prosthetic knee must be taken into account. The proposed design should be able to increase the knee flexion speed without changing or modifying the actuated motor.

1.4. Objectives of the Project

The objectives that are set to be achieved at the end of this project are as follows:

1. To improve the initial design of the microcontroller prosthetic knee in terms of its functionality, size, cost, and weight.
2. To evaluate the strength and range of motion of the proposed design of the microcontroller prosthetic knee.

1.5. Project Schedule

The time schedule of the thesis project is given in Table 1.2.

Table 1.2: The time schedule of the project

Week / Activities	1	2	3	4	5	6	7	8	9	10	11	12	13	14
Determination of thesis title	✓													
Observation, research and collecting information	✓	✓	✓											
Identifying the problem statement		✓												
List down the objectives to be fulfilled			✓											
List down the design criteria			✓											
Rough idea for the design			✓	✓	✓									
Finalize and modelling of the design						✓	✓	✓	✓					
Simulation and analysis of the design										✓	✓	✓		
Report writing													✓	✓

1.6. Organisation of the Project

In Chapter 1, *Introduction*, the basic information of the knee joint and lower limb amputation are described. Different types of prosthetic knees are tabulated in a table. Furthermore, the problems of initial design of the prosthetic knee are discussed and the objectives of the project are listed down. The project schedule shows the management of this project throughout the 14 weeks.

In Chapter 2, *Literature Review*, the several studies in transfemoral prostheses are reviewed. A few different types of commercial microcontroller prosthetic knees are also mentioned.

In Chapter 3, *Methodology*, the steps taken for this project are explained. All the steps are planned properly before the project get started. The sequences of the steps are summarized in the flowcharts for better understanding.

In Chapter 4, *Result*, the proposed design of the prosthetic knee is presented. Each component of the design is elaborated based on their functions and properties. The successful of the simulation models of FEA with different materials are shown. The estimated weight and fabrication cost of the design are calculated.

In Chapter 5, *Discussion*, the further explanation of the design and the simulation models are discussed. The future works for improvement of the project are suggested.

In Chapter 6, *Conclusion*, the whole project is briefly summarized.

Chapter 2. LITERATURE REVIEW

In the past, there were not many devices available for people who lost their lower limb. Most of the time, they use wheelchairs, walkers, and crutches to perform their daily activities. Development of knee prostheses has been taking place for decades. It started with purely mechanical systems to systems that use microprocessor control. Nowadays, the amputees can take advantage in technology and medical science by wearing microcontroller-based lower limb prosthetics. Many efforts have been made to develop a microprocessor transfemoral prosthesis in order to allow the amputees to have a better quality of life. There are different types of actuation methods have been introduced, including electric motors, hydraulic, and pneumatics.

2.1. Research in Transfemoral Prostheses

Many studies in transfemoral prostheses have been carried out for many years. It started in 1970s, where a tethered electrohydraulic transfemoral prosthesis became a pioneer of active joints in transfemoral prosthesis. The design was introduced by Flowers and Mann. It consisted of a hydraulic actuated knee joint tethered to a hydraulic power source and off-board electronics, and computation as shown in Figure 2.1. An ‘echo control’ scheme for gait control was developed where a modified knee trajectory from the sound leg was played back on the contralateral side. The device was designed for able bodied subjects and it could not be tested on an amputee. The device was attached to the able bodied subjects, with their knees in extreme flexion. The subjects had difficulty to walk due to unnatural configuration of the leg (Flowers & Mann, 1977).

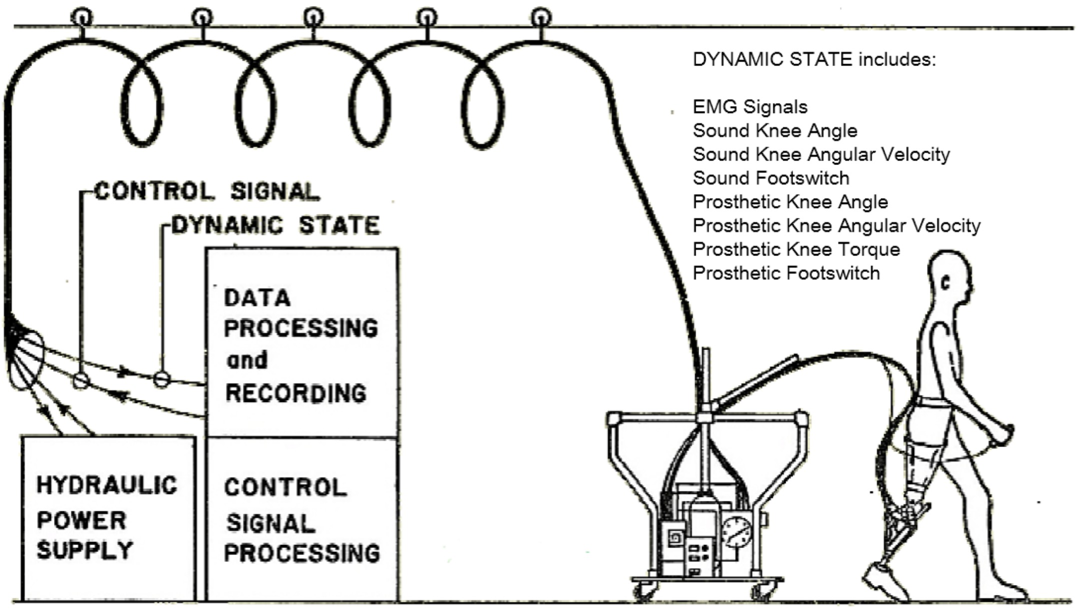


Figure 2.1: The hydraulically actuated knee prosthesis that pioneered the use of active joints (Reproduced from Flowers & Mann, 1977)

At the beginning of the 21st century, a study of direct actuation of the joint by an AC servo motor was introduced by Hata. A complex model to simulate human walking gait was developed together with a few different stiffness mechanism to minimize the torque required when the leg was in a static position. The design was proven to be able to provide roughly 100 Nm of torque and 6.5 rad sec^{-1} of a rotational velocity. Unfortunately, the design required a DC/AC inverter to make the design portable (Hata, 2002).

In 2007, a powered prosthesis using a DC electric motor was designed by Fite et al. Instead of actuating the knee joint directly, the motor turned a ball screw in slider-crank mechanisms. Without using heavy gearboxes, the design allowed the motor to generate a large joint torque. A combination of 150 W DC motor and ball screw was able to provide an output force of 1880 N. The design had a capability to provide torque for walking and stair climbing for users up to 85 kg. The mass of the prosthesis was approximately 3 kg, and it must be tethered to a workstation to provide power and control. A large quantity of batteries was required to operate the prosthesis, which made the system difficult to use (Fite, 2007).

The usage of actuated motors was continued where a series of elastic actuators in an agonist-antagonist configuration was designed by Martinez and Herr. They called the design as agonist-antagonist active knee prosthesis (AAAKP). The knee joint was controlled by the two actuators together as shown in Figure 2.2, and the forces of the actuators could be varied to adjust the stiffness of the joint. The design consisted of a motor lead screw combination, which connected to a linear spring. This design allowed a good control of the knee joint, but the actuator was fairly heavy with a weight of 1.13 kg. It also had a small output force (750 N), where it required a large amount of moment arm to produce enough torque as needed by the knee joint (Martinez et al., 2008).

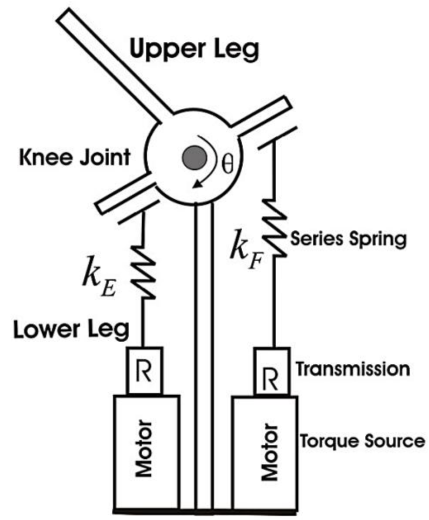


Figure 2.2: Simplified mechanical of the agonist-antagonist active knee prosthesis (Reproduced from Martinez et al., 2008)

The AAAKP inspired by the muscle anatomy of the natural human knee joint. The prototype of the AAAKP is shown in Figure 2.3. The knee joint was controlled by a series of cables which connected to a linear carriage, which free to move along the length of the device. The carriage could be moved on either side depending on the flexion or extension of the springs which driven by an electric motor. An aluminium structure was selected to support the actuation mechanisms, which resembled the shape of the knee and shin. A brushed DC motor was selected in providing sufficient power for both level ground walking, and more extreme tasks, such as stair climbing (Martinez et al., 2008).

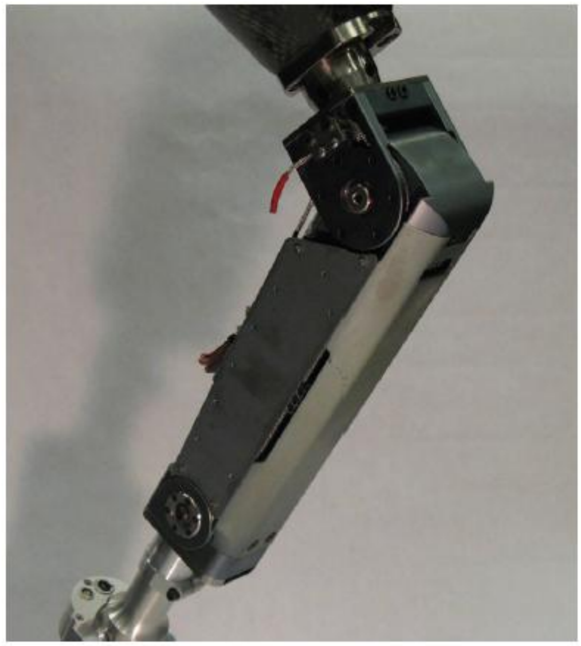


Figure 2.3: The active agonist-antagonist knee prosthesis (Reproduced from Martinez et al., 2008)

Instead of an actuated motor that has been used by many studies, a pneumatic cylinder was introduced by Sup et al. to act as a prosthetic actuator. The design consisted of a powered prosthesis with actuated knee and ankle joints. The actuators were placed in a slider-crank configuration, similar to the slider-crank by Fite (2007). The design could generate a torque curve for a 75 kg user. Since pneumatic actuators were used, the joint compliance was controlled by adjusting the pressure in the cylinder. The authors proposed to use a chemo-fluidic actuation (monopropellant and catalyst) to produce pressurized gas, which is more energy dense (Sup, Bohara, & Goldfarb, 2008).

In 2009, Sup et al. once again produced another powered prosthesis. The initial design was improved by replacing the pneumatic cylinder with a motor-ball screw actuators. The design is illustrated in Figure 2.4. A compression spring was added into the design of the ankle actuator, which similar to Martinez and Herr. The spring helped in generating additional force to be produced at toe-off of the foot. The new design has a mass of 4.2 kg. It could support up to 85 kg of a user over a distance of 9 km (Sup et al., 2009).

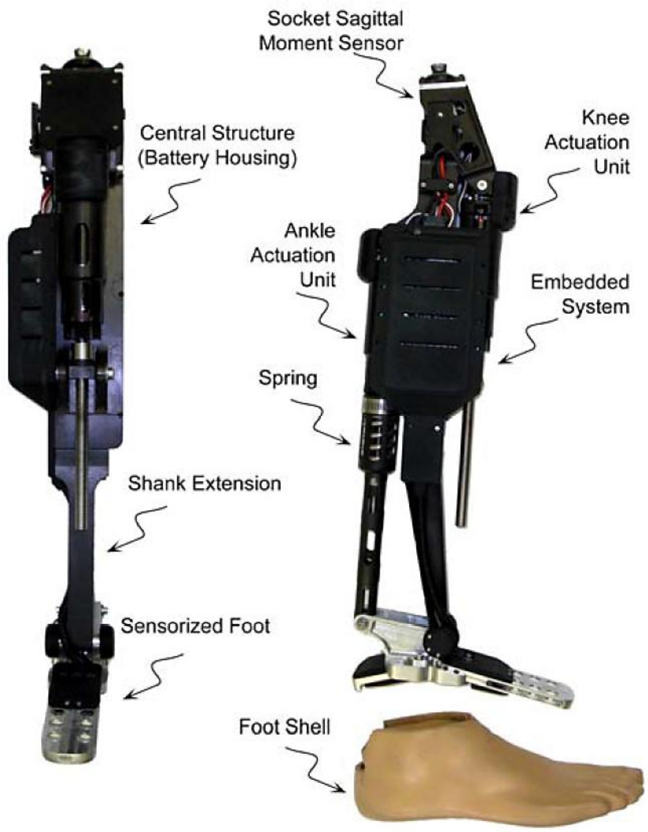


Figure 2.4: The self-contained powered knee and ankle transfemoral prosthesis (front and side views)

(Reproduced from Sup, 2009)

In the same year of Sup et al. improving their design, Lambrecht and Kazerooni developed a semi active hydraulic prosthesis similar to commercial products, such as C-Leg® and

Rheo Knee ®. The design played a role during large portions of the gait cycle by adjusting the joint dumping to improve swing control. A hydraulic pump assisted the knee flexion and extension to ensure a normal gait cycle was maintained. It provided short term power for different types of daily activities, such as standing and stair climbing. The mass of the prosthesis was less than 4 kg. It used a combination of active and passive modes which enable users to walk up to 3000 steps per day. The prosthesis could not contribute to a large amount of positive work during walking. It was designed mainly to assist a user with a powered swing and stance (Lambrecht & Kazerooni, 2009).

The hydraulic knee prosthesis (by Lambrecht and Kazerooni) was assembled from several components as illustrated Figure 2.5. A residual limb of an amputee would be fit onto a conventional prosthetic socket, which attached to the thigh link of the knee. The thigh was hinged to a shank like, where it was connected to an ankle foot prosthesis by a conventional pylon. The hydraulic manifold (contains a hydraulic actuator) connected the thigh and shank links to complete the linkage. The gear pump (acted as a drive motor), a control valve with fluid passages, passive elements, and sensors were connected by the hydraulic manifold. The drive motor, pump displacement, manifold passageways, cylinder size, and linkage geometry were designed carefully to achieve the desired performance (Lambrecht & Kazerooni, 2009)

The development of the transfemoral knee prostheses are briefly explained above. In general, every study showed different designs and mechanisms. Table 2.1 shows the mechanical characteristic from some of studies of transfemoral knee prostheses.

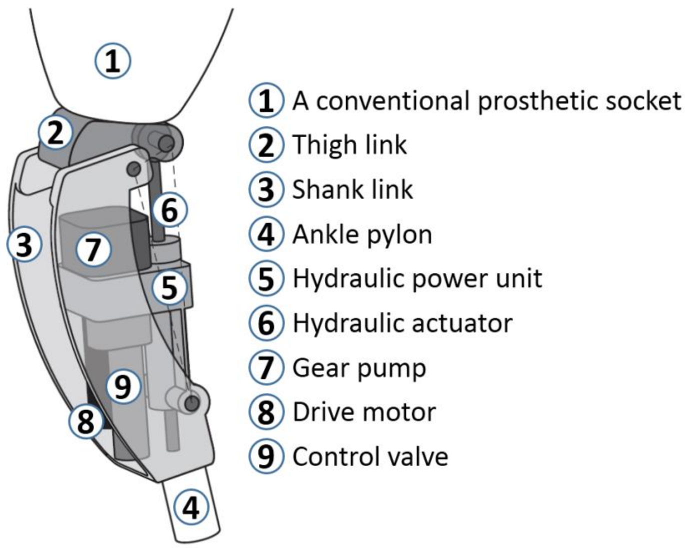
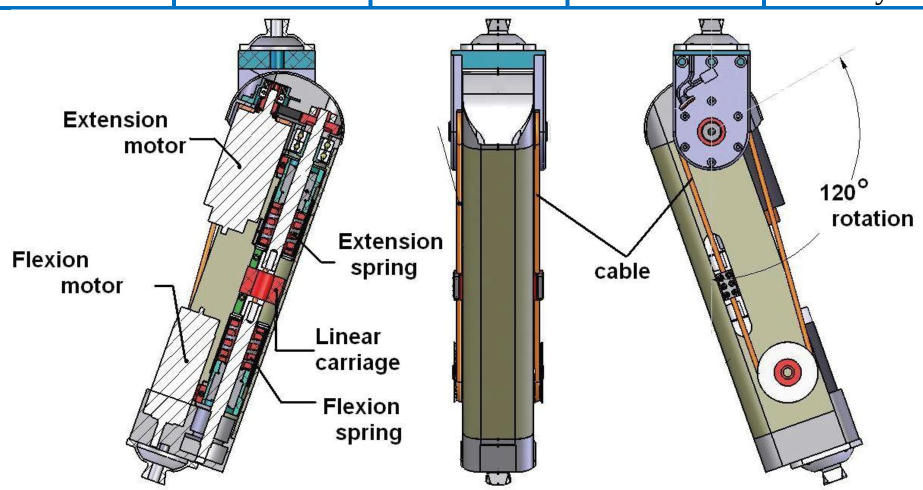


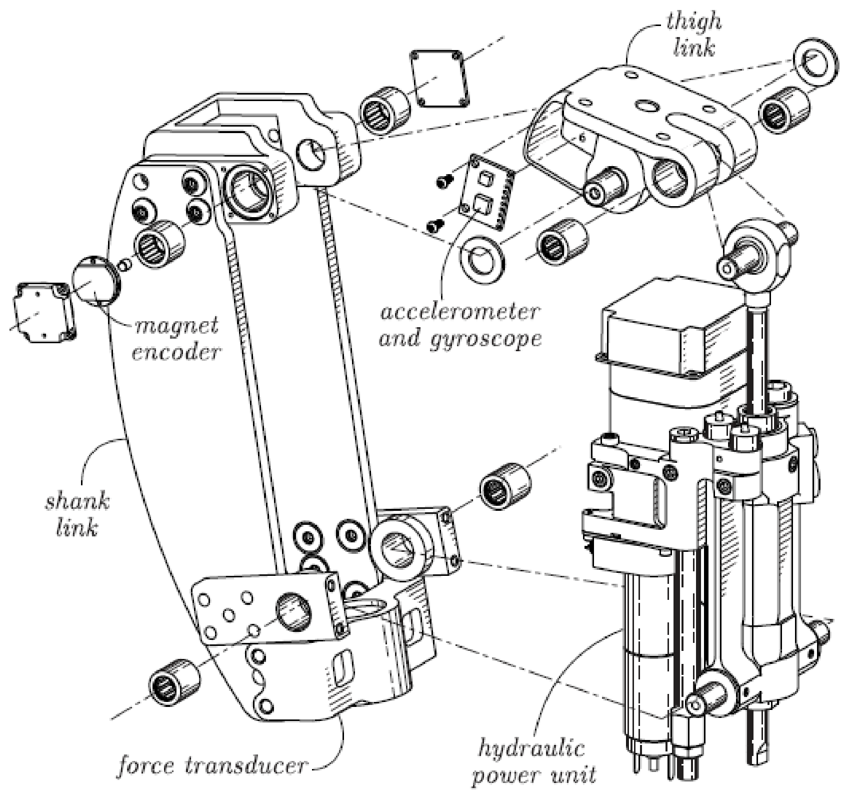
Figure 2.5: Semi-active knee components (Reproduced from Lambrecht and Kazerooni, 2009)

Table 2.1: Summary of some previous works in transfemoral prostheses

<table>
<tr><td>Study 1</td><td colspan="5">Mechanism</td></tr>
<tr><td rowspan="4">Martinez et al. (2008)</td><td colspan="5">Driven by a set of cables with two motors (extension and flexion)</td></tr>
<tr><td>Range of motion</td><td>Material used</td><td>Weight</td><td>Dimension</td><td>Testing</td></tr>
<tr><td>$0 – 120^{O}$</td><td>Aluminium</td><td>Unknown</td><td>Unknown</td><td>Simulation only</td></tr>
<tr><td colspan="5">

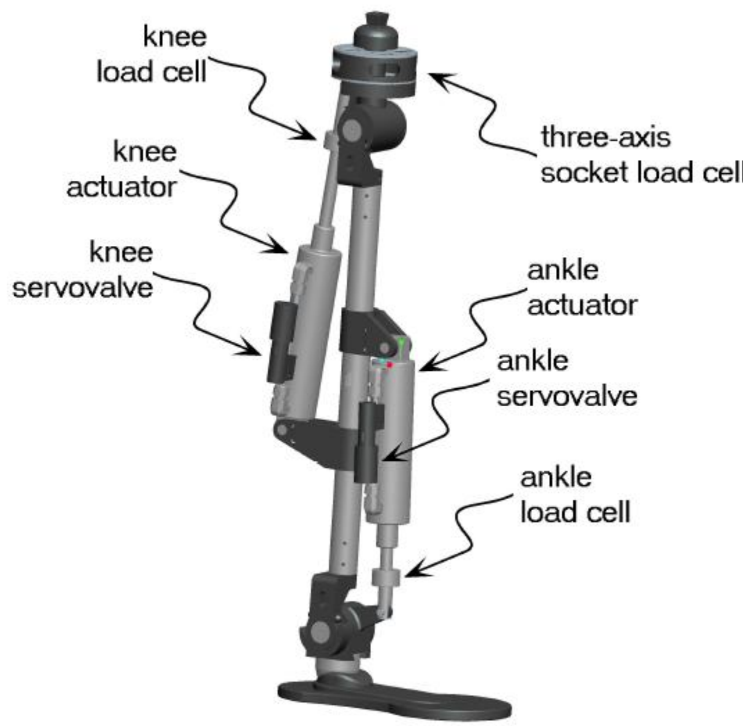
(Reproduced from Martinez et al., 2008)</td></tr>
</table>

Table 2.1, continued

<table>
<tr><td>Study 2</td><td colspan="5">Mechanism</td></tr>
<tr><td rowspan="4">Lambrecht (2008)</td><td colspan="5">Uses hydraulic manifold to flex the knee</td></tr>
<tr><td>Range of motion</td><td>Material used</td><td>Weight</td><td>Dimension</td><td>Testing</td></tr>
<tr><td>0 – 120^{O}</td><td>Carbon fiber (frame), aluminium (holders)</td><td>Unknown</td><td>Unknown</td><td>Tested on amputee</td></tr>
<tr><td colspan="5">

(Reproduced from Lambrecth, 2008)</td></tr>
</table>

Table 2.1, continued

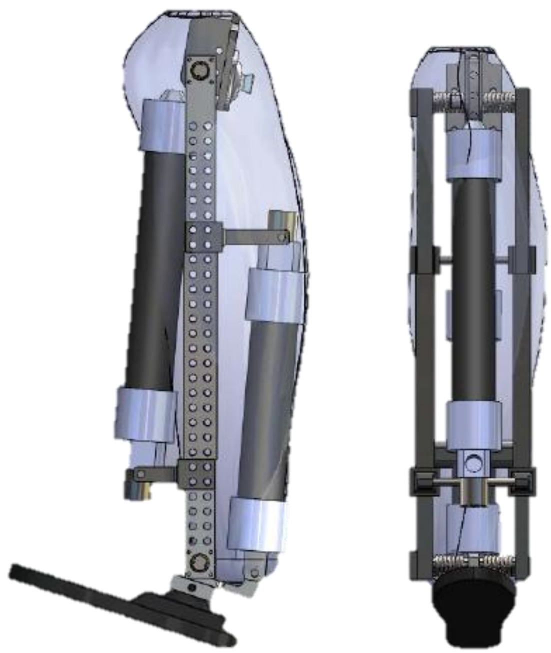
<table>
<tr><td>Study 3</td><td colspan="5">Mechanism</td></tr>
<tr><td rowspan="4">Sup
(2008)</td><td colspan="5">Uses double-acting pneumatic actuators (ankle and knee)</td></tr>
<tr><td>Range of motion</td><td>Material used</td><td>Weight</td><td>Dimension</td><td>Testing</td></tr>
<tr><td>0 – 110^{O}</td><td>7075-T6 Aluminium</td><td>Unknown</td><td>Unknown</td><td>Tested on a healthy subject</td></tr>
<tr><td colspan="5">

(Reproduced from Sup et al., 2008)</td></tr>
</table>

Table 2.1, continued

<table>
<tr><td>Study 4</td><td colspan="5">Mechanism</td></tr>
<tr><td rowspan="4">Fite
(2007)</td><td colspan="5">Uses a 150 W brushed DC motor (Maxon model RE40)</td></tr>
<tr><td>Range of motion</td><td>Material used</td><td>Weight</td><td>Dimension</td><td>Testing</td></tr>
<tr><td>Unknown</td><td>Unknown</td><td>3.05 Kg</td><td>Unknown</td><td>Tested on a healthy person</td></tr>
<tr><td colspan="5">
(Reproduced from Fite, 2007)</td></tr>
</table>

Table 2.1, continued

<table>
<tr><th>Study 5</th><th colspan="5">Mechanism</th></tr>
<tr><td rowspan="4">Waycaster (2010)</td><td colspan="5">Uses two actuators (knee and ankle) and torsional spring for easier adjustability of the leg length.</td></tr>
<tr><td>Range of motion</td><td>Material used</td><td>Weight</td><td>Dimension</td><td>Testing</td></tr>
<tr><td>$0 – 110^{O}$</td><td>6061 T6 Aluminium alloy</td><td>Unknown</td><td>Unknown</td><td>Tested on a healthy subject</td></tr>
<tr><td colspan="5">
(Reproduced from Waycaster, 2010)</td></tr>
</table>

2.2. *Commercial Microcontroller Prosthetic Knees*

Several different types of prosthetic knees are commercially available today. They range from simple mechanical joints to advanced microcontroller joints. Each one of them has its own strength and weaknesses based on the activity level and condition of the user. This section describes several types of commercial microcontroller prosthetic knees and their working method.

Some of the most commercially successful of microcontroller prosthetic knees are the OttoBock C-Leg® (OttoBock, 2014), the Ossur RHEO Knee® (Ossur, 2014), and the Freedom Innovations Plié Knee® (Innovations, 2014). They use a hydraulic joint that is controlled by a microcontroller. They allow some variable damping throughout the swing phase. They also have the ability to lock the knee joint in any position. The joint angle can be determined by using a potentiometer. Then, based on several programmes of walking modes, a stepper motor will automatically adjust~~s~~ the size of an orifice in a hydraulic cylinder. This allows for dynamically variable damping, which can be adjusted depending on different types of activities.

In 1997, the OttoBock C-Leg® was first introduced to the public and it made a history with the first microprocessor controlled knee joint in the world. It has capabilities to regulate not just the swing phase control, but also the locking during stance phase. Two sensors at the knee and ankle supply information to a microprocessor which placed in the shank of the prosthesis. Based on this information, the microprocessor controls the hydraulic in real time. The energy needed to perform this operation is supplied by an integrated battery. The sensors in the OttoBock C-Leg® replace some of the sensors in the body. Meanwhile, the hydraulic stabilizes some of the muscle functions in the body. During stance phase, the main task is stabilizing the leg so that it can support the body weight. During the swing phase, a dynamic controlled is required where it controls the leg as it swings through the lower leg, and it slows down in preparation for the next step. The basic working principles of the OttoBock C-Leg® are illustrated in Figure 2.6.

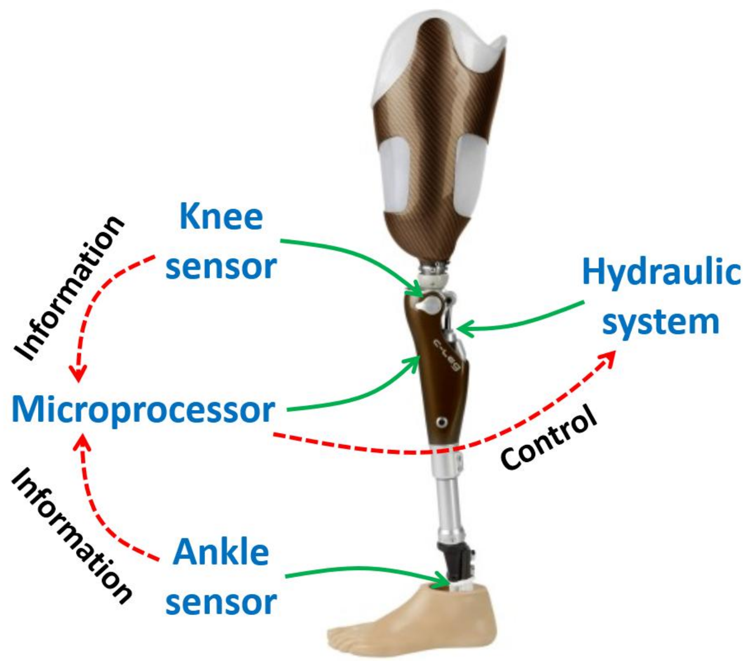


Figure 2.6: How the OttoBock C-Leg® works

The Ossur RHEO Knee® uses an electric motor to provide power at the knee joint. The picture of the Ossur RHEO Knee® is shown in Figure 2.7. It assists amputees during both phases of gait cycle, and allows different types of activities such as stair climbing and standing from sitting position. It uses an echo control from a user's sound leg, which means bilateral amputees are not able to use the Ossur RHEO Knee®. It weighs about 4.5 kg, approximately three times the mass of the OttoBock C-Leg®. It costs about USD 120,000 (5-6 times more than the OttoBock C-Leg®).

Figure 2.7: The Ossur RHEO Knee®

As discussed above, the designs of the prosthetic knees in the market varies greatly amongst them. The classification and comparison of input method, control, and actuation for currently available motor-actuated prosthetic knees are shown in Table 2.2.

Table 2.2: The classification and comparion of input method, control, and actuation for some of the prosthetic knees (Reproduced from Martin, 2010)

Device	Input method	Sensor input	Control methods	Actuation method	Additional unique control features
Knee					
Otto Bock (Duderstadt, Germany) C-Leg	CIC	Knee angle sensor, force-sensing strain gauge, time counter	Real-time control and variably resistive in stance and swing	Hydraulic valve actuator	Hydraulic valve actuators have inherent hydrodynamics that account for terrain and speed changes within a narrow range, and microprocessor-controlled valves broaden this range. Practitioner programmable settings

Table 2.2, continued

Device	Input method	Sensor input	Control methods	Actuation method	Additional unique control features
Otto Bock Compact	CIC	Knee angle sensor, force-sensing strain gauge, time counter	Real-time control and variably resistive in stance, but not in swing	Hydraulic valve actuator	Controls for a limited gait cadence. Practitioner programmable settings
Ossur (Reykjavik, Iceland) Rheo	CIC	Angle sensor, force-sensing stain gauge, time counter	Real-time control and variably resistive in stance and swing	Magnetorheological actuator	Self learning program. Practitioner programmable settings
Ossur Power Knee	CIC, with using IEC sound foot inputs	Sound Limb gyrometers, pressure and load cells to measure motion, position, and velocity, bluetooth transfer, angle sensor, time counter	Real-time control and variably powered in stance and swing	Electromechanical actuator	Replaces lost concentric muscle function to restore symmetry, balance, and power. Offers spatial orientation of the limb. Practitioner programmable settings
Freedom Innovations (Irvine, CA) Plié MPC	CIC	Extension sensor, force-sensing strain gauges, time counter, step counter	Microprocessor management of stance-swing transition, hydraulic control of resistance during swing	Hydraulic valve actuator	Practitioner programmable settings
Endolite (Centerville, OH) IP Plus	CIC	Angle sensor, ambulation speed sensor, time counter	Variable resistive actuation in swing, mechanical breaking weight-activated stance	Pneumatic piston	Practitioner programmable to specific activities, cushion valve, energy management
Endolite Smart IP	CIC	Angle sensor, ambulation speed sensor, time counter	Variable resistive actuation in swing, mechanical breaking weight-activated stance	Pneumatic piston	Practitioner programmable to specific activities, self learning
Endolite Smart Adaptive	CIC	Angle sensor, force sensor, time counter	Variable resistive actuation in swing and stance	Swing–pneumatic piston, stance–hydraulic actuator	Practitioner programmable to specific activities, automatic modes for stairs, stance, stumble recovery, standing, accommodates ambulation speed and slopes

Chapter 3. METHODOLOGY

3.1. Introduction

This thesis project was completed in one semester. The flow diagram in Figure 3.1 shows the stages to be taken into consideration before this project is completed.

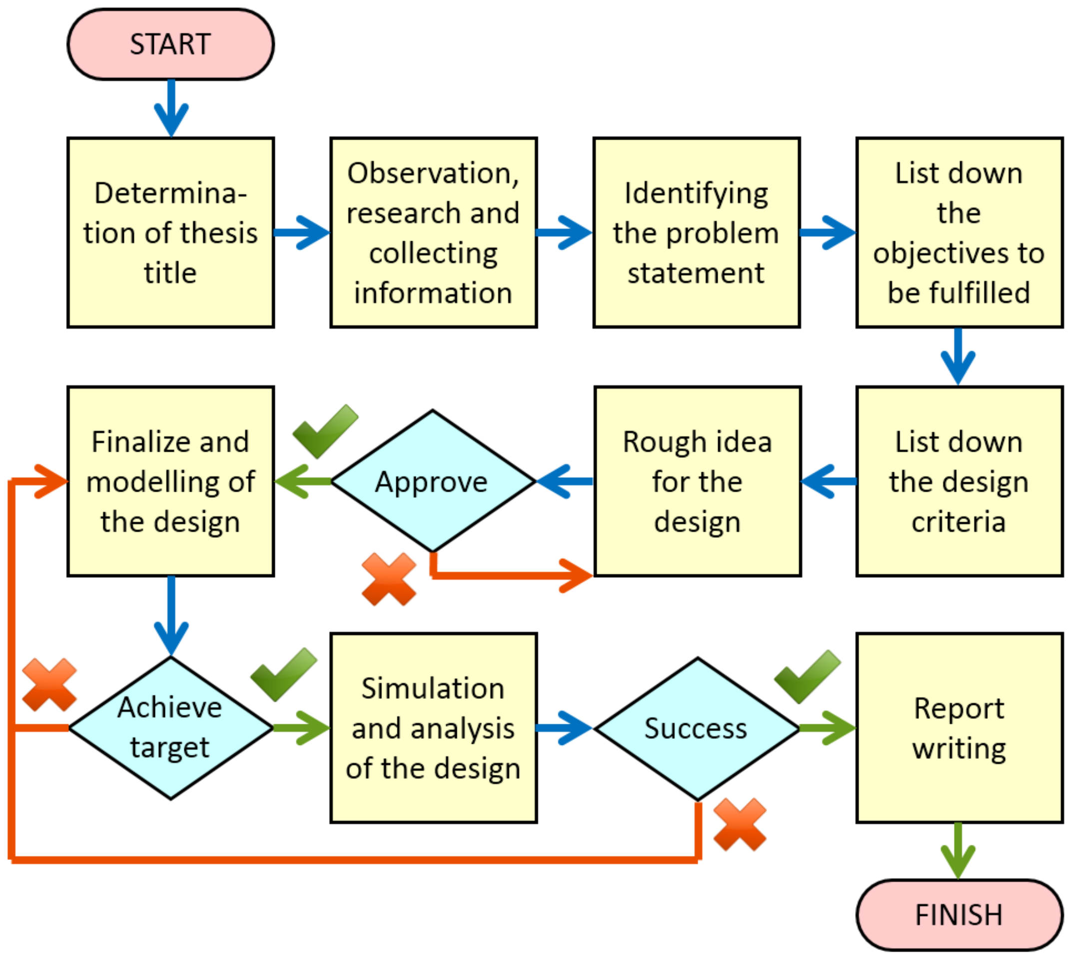


Figure 3.1: Flow diagram for the thesis project

3.2. Observation, Research, and Collecting Information

Getting enough information is crucial to ensure the smoothness of this project. The first step was taken by searching the information related to prosthetic knees from different sources such as journals and books. Several studies of prosthetic knees were observed and compared. At the same time, a few commercially successful microprocessor prosthetic knees were studied.

3.2.1. First Prototype Analysis

All information about the first prototype, i.e. the initial design was collected by observing the first prototype that is available in the laboratory. A few photographs were captured during the visit as shown in Figure 3.2. Meanwhile, Figure 3.3 shows each component of the first prototype.

Table 3.1 shows the collected information from the first prototype of the microcontroller prosthetic knee. Martinez et al. (2008) had listed down a few reasonable design criteria of a prosthetic knee, which was shown at the top-right corner of the Table 3.1. The dimension, weight, and flexion angle range were compared between both the initial design and the reasonable design criteria. In terms of size, the initial design can be considered as bulky, and unsuitable to be worn by amputees who have short lower limbs. The weight of approximately 3 Kg fulfils the reasonable design criteria. The flexion angle range of the initial design can only go up to 90^{O} instead of 120^{O}. In order to mimic the function of a normal knee, the prosthetic knee must be able to flex more than 90^{O} (Kingston, 2001).

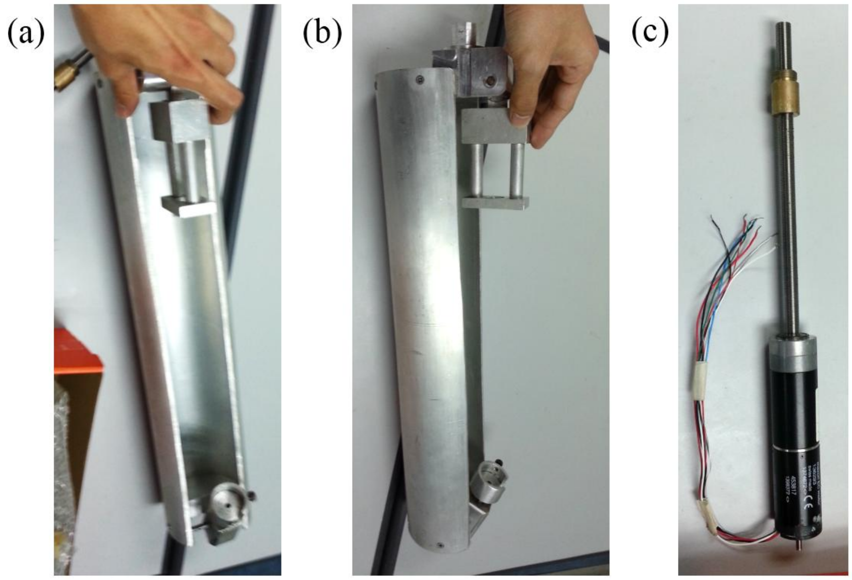


Figure 3.2: The photographs of the first prototype, where (a) and (b) were taken from the posterior and lateral view respectively. Photograph (c) is the motor used as the actuator

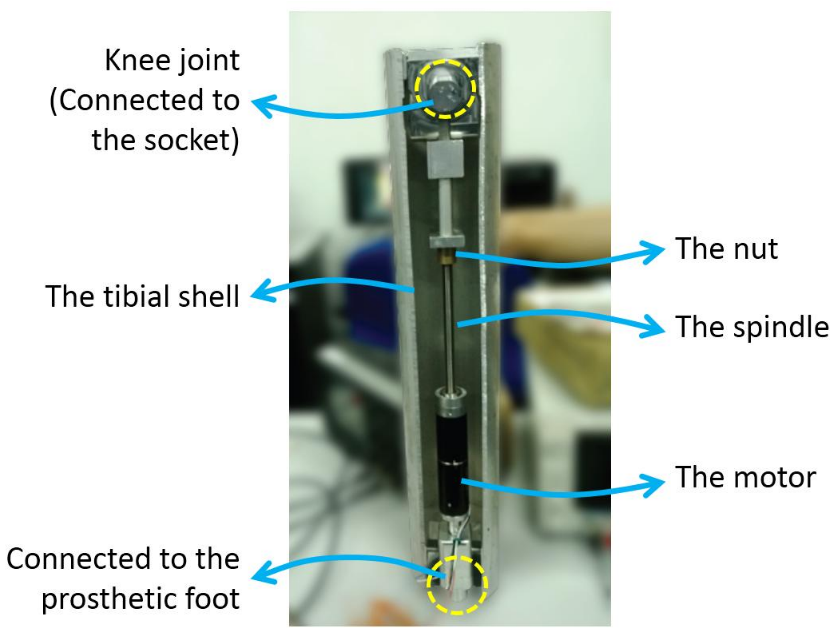


Figure 3.3: The components of the first prototype

Table 3.1: Analysis of the first prototype.

<table>
<tr><td></td><td></td><td>Reasonable design criteria of a prosthetic knee.
(Adapted from Martinez et al., 2008).</td></tr>
<tr><td>Dimension</td><td>Length: 450 mm
Width: 110 mm ✖</td><td>Length: 330 mm
Width: 70 mm</td></tr>
<tr><td>Weight</td><td>~ 3 Kg ✔</td><td>3 Kg</td></tr>
<tr><td>Flexion angle range</td><td>0^{O} - 90^{O} ✖</td><td>0^{O} - 120^{O}</td></tr>
<tr><td>Material used</td><td colspan="2">Aluminium 6061</td></tr>
<tr><td>Knee joint type</td><td colspan="2">Simple, hinge based,
one degree of freedom</td></tr>
<tr><td>Time taken to flex the knee up to 90^{O}</td><td colspan="2">~ 40 seconds (2.25 deg/s) ✖</td></tr>
<tr><td>Gearing structure</td><td colspan="2">No gearing structure.</td></tr>
<tr><td>Aesthetic value</td><td colspan="2">A very long cylinder object (looks like a bazooka) ✖</td></tr>
<tr><td rowspan="8">Motor used</td><td>Brand</td><td>Maxon spindle drive</td></tr>
<tr><td>Model of the spindle</td><td>Spindle drive GP 32 S,
Ø 32 mm,
Metric spindle</td></tr>
<tr><td>Length of the spindle</td><td>200 mm</td></tr>
<tr><td>Spindle pitch to pitch value</td><td>1 mm</td></tr>
<tr><td>Max output speed</td><td>Theoretical: 25000 RPM
Practical: 16700 RPM</td></tr>
<tr><td>Gear ratio</td><td>492:1</td></tr>
<tr><td>Speed of the spindle</td><td>Theoretical: 50.8 RPM
Practical: 33.9 RPM</td></tr>
<tr><td>Maximum feed velocity</td><td>Practical: 0.5 mm/s ✖</td></tr>
</table>

Note: ✔ OK ✖ NOT OK

The initial design was fabricated using aluminium 6061 thus it is lightweight. The knee joint consists of a simple, hinged based structure. It allows a movement of one degree of freedom. The most obvious issue of the initial design is the time taken for the knee to flex. It takes approximately 40 seconds to flex the knee from 0 to 90^{O}, which is too slow and unacceptable for practical application. The initial design contains no gearing structure. The long cylindrical shell of the initial design does not look appealing and a new cover to increase the aesthetic value must be done.

A Maxon spindle drive motor was purchased to be used in this project. The model of the spindle is Spindle drive GP 32 S (metric spindle) with a diameter of 32 mm. It has a length of 200 mm, and a pitch to pitch value of 1 mm. For the actuated motor, the maximum output speed is 25000 RPM, which is quite fast. However, due to some limitations with the power supply in the laboratory, the maximum speed reaches up to 16700 RPM only. A gear ratio of 492:1 makes the actuated motor to have high torque, so that it has enough strength to flex the prosthetic knee. After assembling the actuated motor to the initial design, it proved that the actuated motor has enough strength to flex the prosthetic knee up to 90^{O}. But, in a very slow motion. The maximum feed velocity of 0.5 mm/s was measured manually using a stopwatch and a ruler. This value contributes to the longer time for the prosthetic knee to flex.

3.3. Identifying the Problem Statement

The initial design was carefully examined. After the information was gathered, the pros and cons of the initial design were determined. The major concern is about the functionality and usability of the initial design. The prosthetic knee of the initial design flexes in a very slow speed. The flexion angle range does not mimic a normal, healthy knee. The initial design is not user-friendly due to its size. It also has low aesthetic value.

3.4. Identification of Objectives

After all the problems were determined, the method to solve them must be found. This project will try to solve those problems which are currently found in the initial design of the prosthetic knee. The objectives must be listed down to ensure that this project ends with success. The objectives serve as targets or guidelines to achieve for the final product and the successful implementation of the final design.

3.5. Identification of the Design Criteria

After all problems with the initial design were defined, the problems must be translated into design criteria. These design criteria are the explicit goals in this project that must be achieved in order to be successful. The design criteria of this project are listed in Figure 3.4.

DC 1: Range of motion

i. The prosthetic knee must be able to flex from 0 to 120 degrees.

DC 2: Speed

i. The time taken to flex the prosthetic knee from 0 to 120 degrees must be within 2 seconds. It means the knee flexion velocity is at 60 deg/sec.

DC 3: Dimension

i. The prosthetic knee must be small enough to be used by Malaysian amputees.
ii. The length and the width of the prosthetic knee should be around 250 mm and 70 mm respectively.

DC 4: Weight

i. The prosthetic knee must be light. The materials used should be picked carefully to avoid excessive weight.
ii. The prosthetic knee should have a weight of maximum 3 Kg.

DC 5: Strength

i. The prosthetic knee should be able to withstand a load of 800 N.
ii. The amputees weighted of 80 Kg and less should be able to wear the prosthetic knee.

DC 6: Cost and quality

i. The materials used must be selected based on their cost and quality.
ii. The aesthetic value must be taken into account during the design process.

Figure 3.4: The design criteria for the project

3.5.1. Range of Motion

The maximum angle of a normal knee can bend is 140^{O} as illustrated in Figure 3.5 (a). In most activities, the knee is not fully bent up to 140^{O}. Instead, the flexion angle range required to accomplish most activities is range only from 0 to 120^{O} as illustrated in Figure

3.5 (b) (Kingston, 2001). The first design criteria were set based on this information. By wearing the prosthetic knee, it is expected that the amputees will not be able to achieve the maximum flexion angle of the normal knee. But, it is anticipated that at least the amputees can be able to perform most of their daily activities by allowing the prosthetic knee to flex up to 120^{O}.

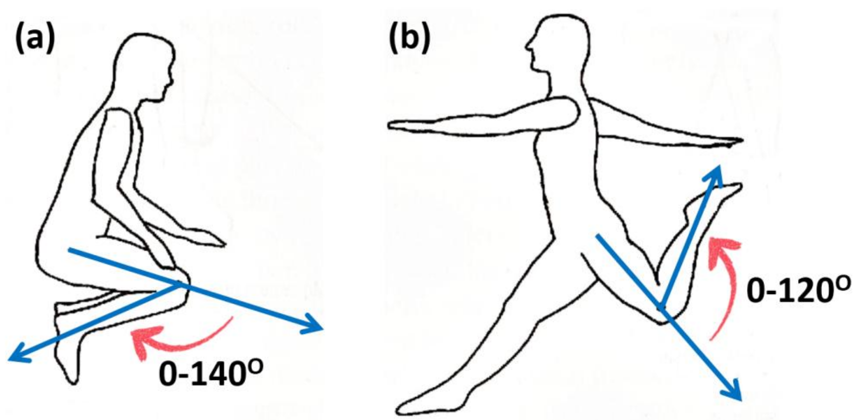


Figure 3.5: Flexion of the knee, (a) The maximum angle of the knee can bend, (b) The angle required for most activities (Adapted from Kingston, 2001)

3.5.2. Speed

Mobility is crucial for human to ensure that they can move from one place to another. The most common activity of human to move is by walking. Figure 3.6 shows graphs of knee flexion angle and knee flexion velocity from a study of 8 normal subjects by Fox and Delp over a range of walking speeds. The study shows that the mean value of the knee flexion velocity in ‘free’ category is at 300 deg/sec. Meanwhile, the mean value in ‘very slow’ category is at 180 deg/sec (Fox & Delp, 2010). The second design criteria was set after considering two things:

i) The walking speed for amputees is slower compared to normal persons (Pitkin, 2010). In that sense, this project aims to have the knee flexion velocity of the prosthetic knee at 60 deg/sec.

ii) Despite the value of 60 deg/sec is still far to mimic the normal knee flexion velocity, it is considered as a big improvement if compared to the knee flexion velocity of 2.25 deg/sec from the initial design of the prosthetic knee.

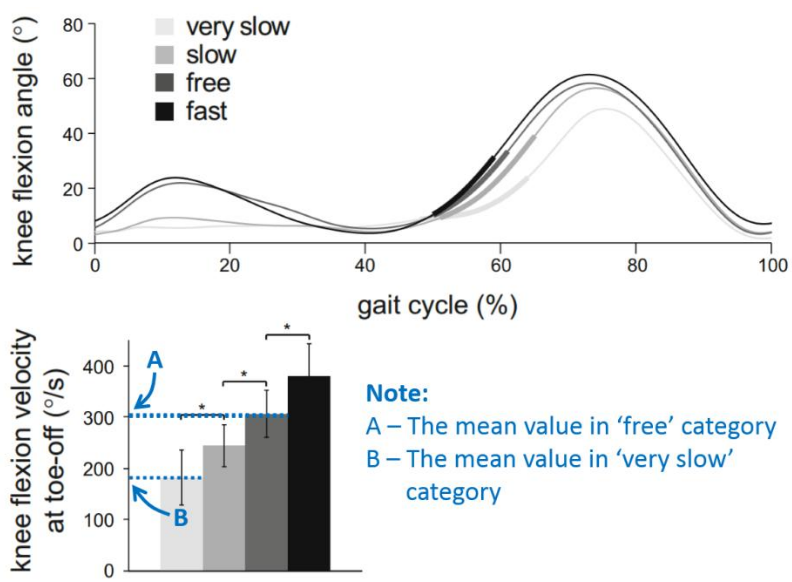


Figure 3.6: The graphs of knee flexion angle and knee flexion velocity in a study of 8 normal subjects (Adapted from Fox & Delp, 2010)

3.5.3. Dimension

The dimension of the prosthetic knee must be small enough to be used by local Malaysian amputees. Table 3.2 shows the mean height and body weight in a national sample of Malaysian adults (Lim & Ding, 2000). The average height of women in ‘All’ category was selected to estimate the minimum length of shank of local Malaysian amputees. From the body anthropometry diagram in Figure 3.7, the shank length that was expressed as a ratio of body height is 0.204.

Table 3.2 - The mean height and body weight in a national sample of Malaysian adults (Adapted from Lim & Ding, 2000)

	All		Malay		Chinese		Indian		Other indigenous	
	M	W	M	W	M	W	M	W	M	W
Height (cm)	165.0	154.0	164.7	153.3	166.9	156.5	167.6	154.7	161.6	150.8
Weight (Kg)	61.8	54.7	61.6	55.0	64.6	55.0	64.9	56.5	58.3	51.5

Note: M – Men, W – Women

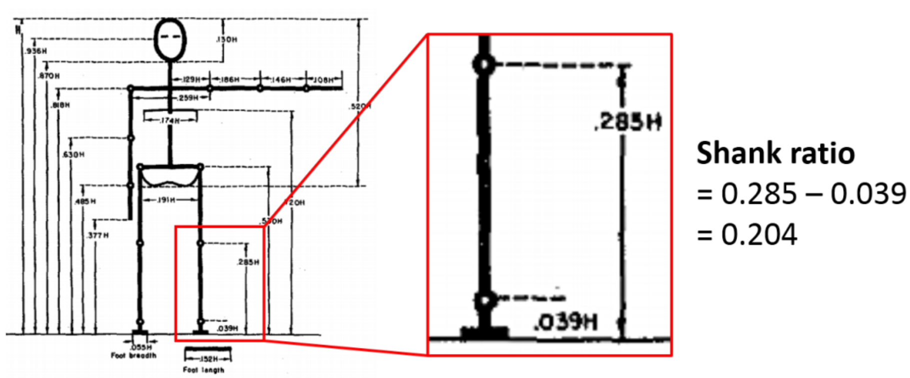


Figure 3.7: The shank length expressed as a ratio of body height (Adapter from Dillis & Contini, 1966)

From the information presented, the estimated minimum length of the shank of local Malaysian amputees is 378.8 mm as illustrated in Figure 3.8. The prosthetic knee unit is expected to have a length of two thirds of the estimated minimum length of the shank. In that case, the expected length of the prosthetic knee unit is approximately 250 mm. The width of the prosthetic knee of 70 mm follows the reasonable design criteria by Martinez et al. (2008).

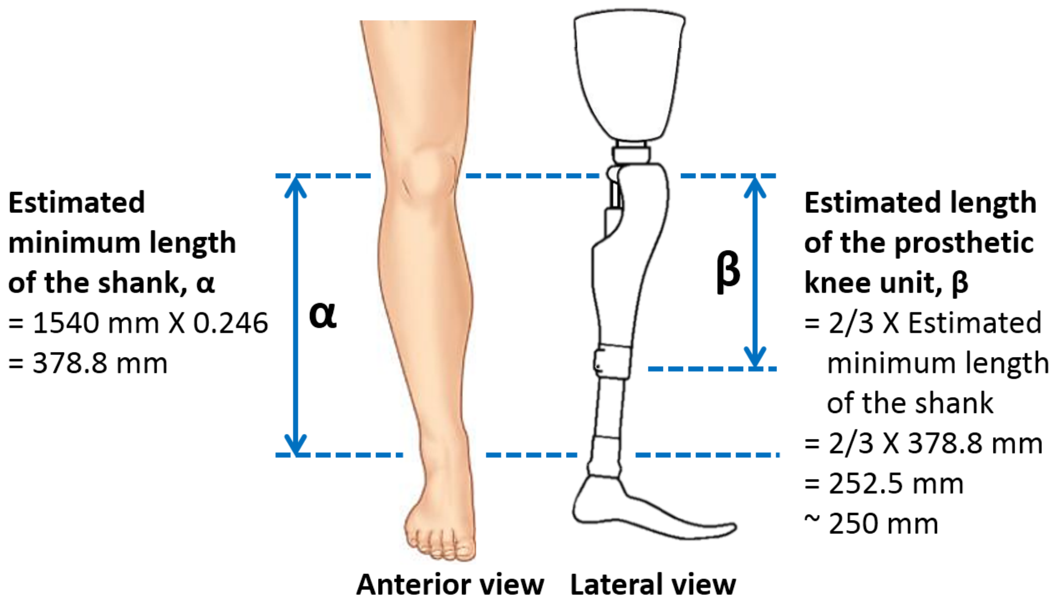


Figure 3.8: The estimated length of the shank, and the estimated length of the prosthetic knee unit

3.5.4. Weight, Strength, Cost and Quality

The weight, strength, cost, and quality relate to materials selection as illustrated in Figure 3.9. The materials must be selected carefully in order to fulfil all four criteria. The weight of 3 Kg was selected based on the reasonable design criteria of a prosthetic knee by

Martinez et al. (2008). By assuming the maximum weight of the amputees is 80 Kg, the strength of the prosthetic knee will be simulated by applying a load of 800 N. The prosthetic knee is not recommended for amputees weighted more than 80 Kg.

A few different materials were selected and simulated. The most favourable material must be selected based on its cost efficiency and quality. The total cost (including the fabrication) should be affordable by most amputees while maintaining the overall quality of the prosthetic knee. The aesthetic value must be taken into account by designing a nice shank cover in order to make the prosthetic knee looks more appealing.

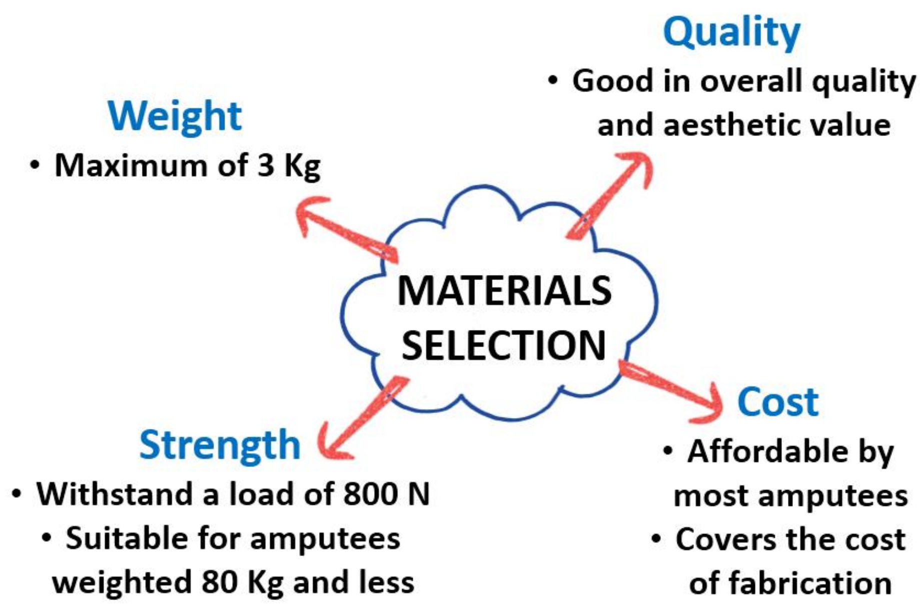


Figure 3.9: Four factors that need to be taken into consideration during materials selection

3.6. Rough idea for the design

After all the information was analysed and the market products were surveyed, the rough idea for the design was formulated. The design criteria were used as guidelines during the design process. The rough idea was drawn and presented to supervisor to obtain advice in terms of mechanical properties and the strength of the design. Two-way communication was practiced with the supervisor for several brainstorming sessions. During the sessions, no idea was judged or discarded. All ideas were recorded for use in the next step of the process.

3.6.1. Rough idea 1

A gearing structure is introduced in the first rough idea to increase the knee flexion angle and knee flexion velocity. An internal gearing structure with a ratio of 10 is used. The pinion and the annular gear are modified to reduce the size as illustrated in Figure 3.10. With a smaller size, they can easily fit in the prosthetic knee.

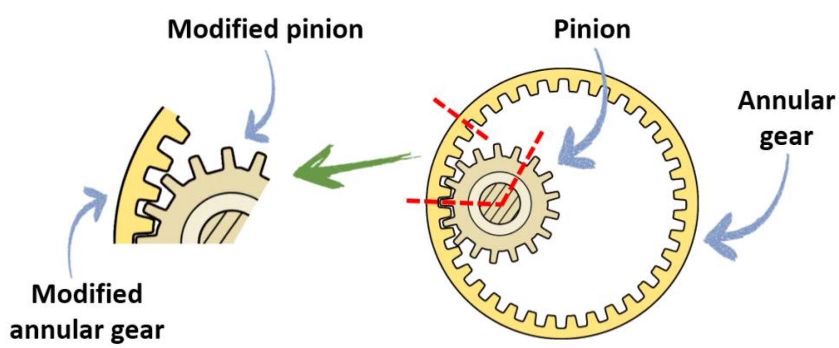


Figure 3.10 : The modified pinion and annular gear of an internal gearing structure (Adapted from Wikipedia)

Figure 3.11 illustrates the first rough idea of the prosthetic knee. The modified annular gear is connected to the nut of the actuated motor. Meanwhile, the modified pinion is connected to the knee joint. The actuated motor is placed at the posterior side of the prosthetic knee as this configuration is commonly found in motorized prosthetic knees in the market. Two long frames on both lateral and medial sides are added to provide strength to the prosthetic knee unit.

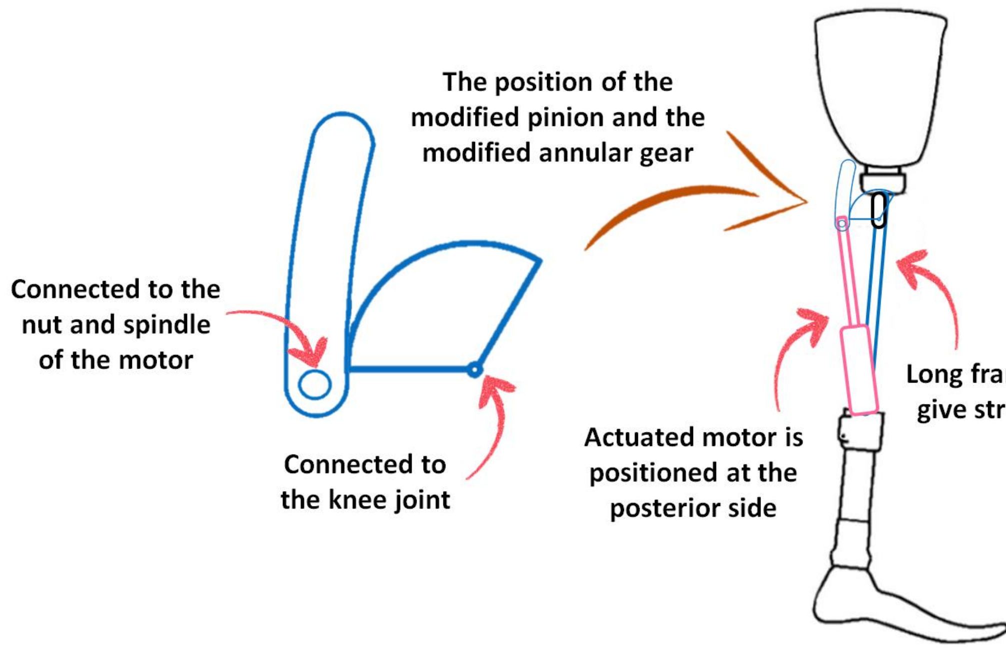


Figure 3.11: The diagram of the first rough idea

Figure 3.12 demonstrates the illustration of the knee flexion of the prosthetic knee at certain degrees. Time reduction of knee flexion from 40 seconds (initial design) to 2 seconds (proposed design) is needed. It means that the proposed design should be able to flex 20 times faster compared to the initial design. As the gear ratio of 10 is used, it means the knee can already flex 10 times faster. But, this is not fast enough. To make the proposed design 20 times faster, the original spindle has to be replaced with a new spindle with a pitch to pitch value of 2 mm as illustrated in Figure 3.13.

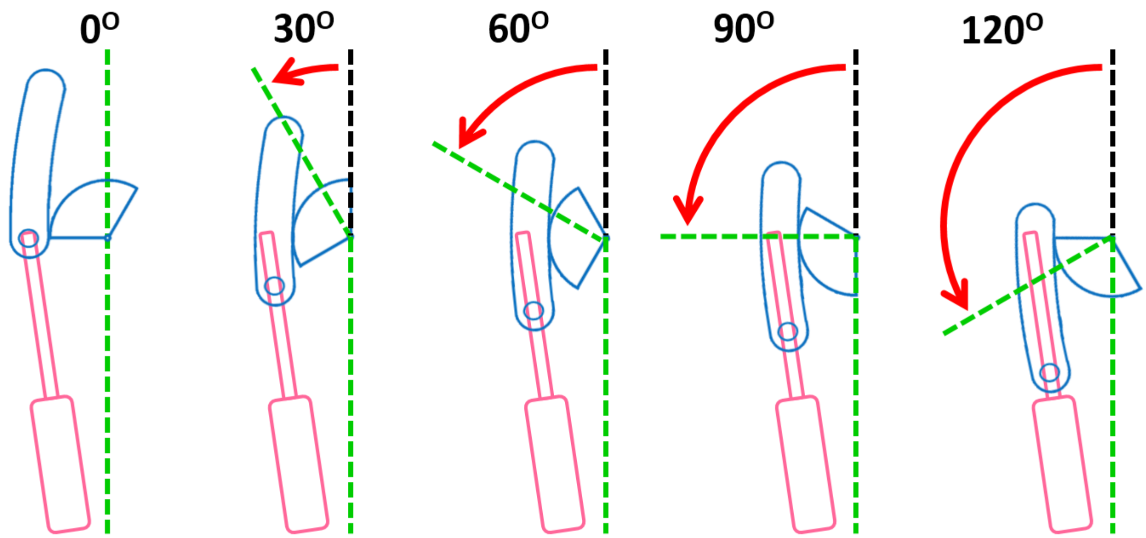


Figure 3.12: The illustration of the knee flexion at certain degrees

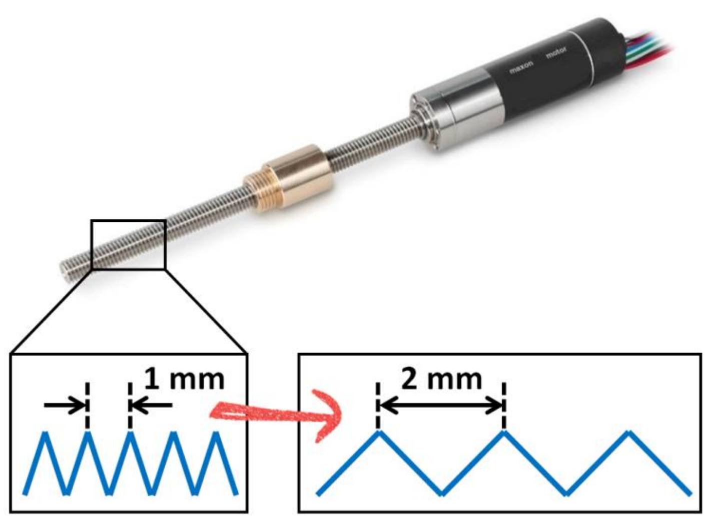


Figure 3.13: The new proposed spindle with 2 mm pitch to pitch value

3.6.2. Rough idea 2

The second rough idea is to use multiple compound gears (same size). These gears will be arranged as a gear train to achieve the goal to have the knee to flex within 2 seconds. The idea is illustrated in Figure 3.14. The actuated motor is placed at the posterior side of the prosthetic knee similar to rough idea 1. The first gear has an extended arm and connected to the nut of the actuated motor. Meanwhile, the last compound gear is positioned at the knee joint. The gear ratio (the first compound gear to the last compound gear) required to flex the prosthetic knee in 2 seconds is illustrated in Figure 3.15.

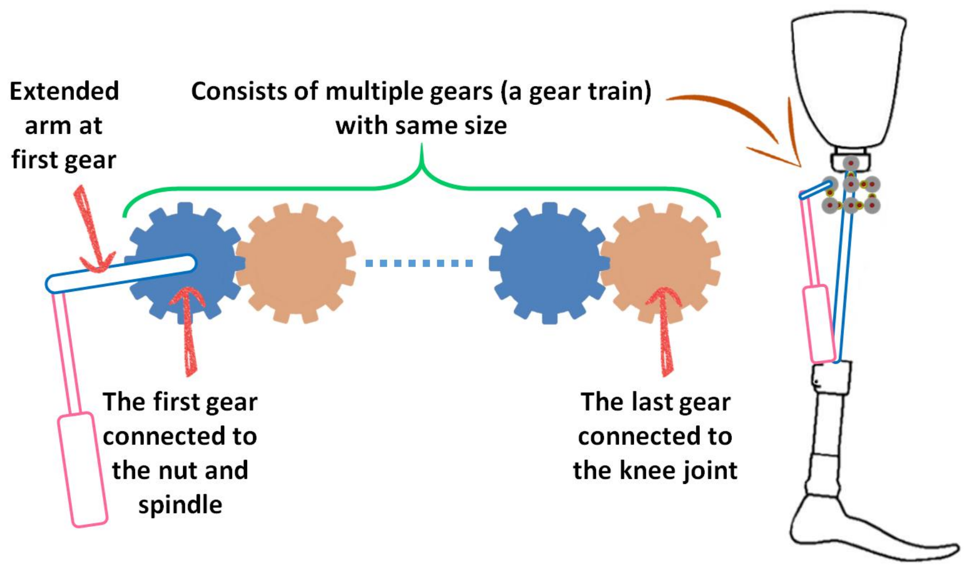


Figure 3.14: The second rough idea

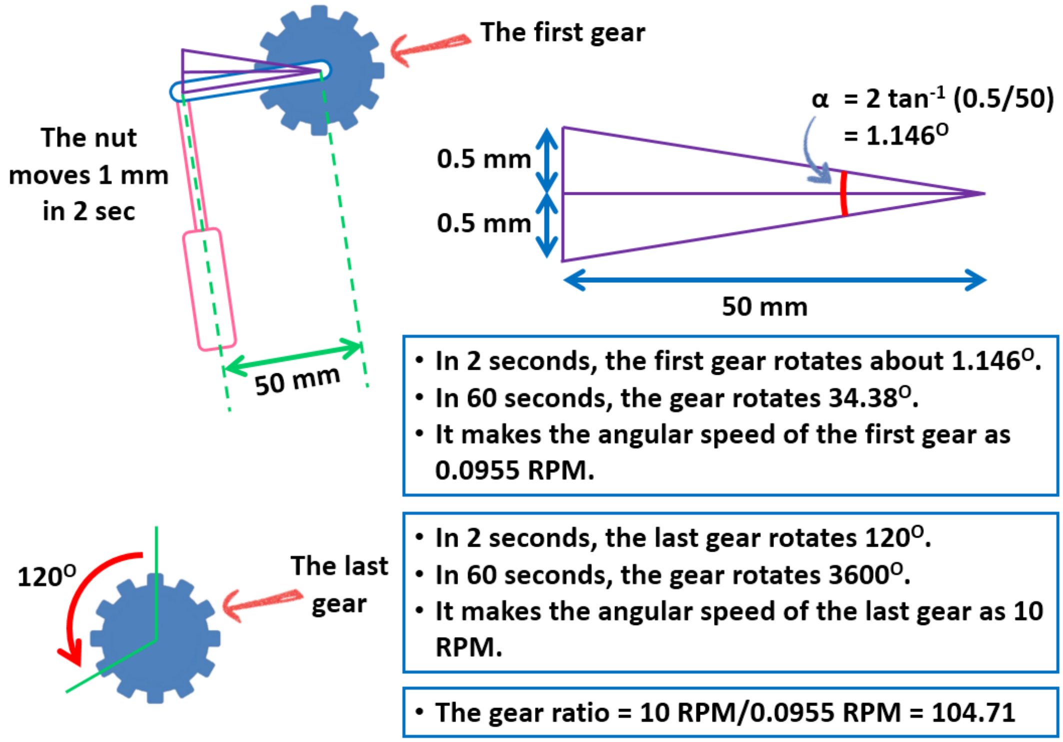


Figure 3.15: Some calculation of the second rough idea

A series of compound gears with a ratio of 2 is selected. From the gear ratio, the total number of compound gears required to get a gear train ratio of 104.71 is 7 gears as calculated below:

$2^n = 104.71$, where n is the number of compound gears needed to flex the prosthetic knee in 2 seconds.

$n = \log 104.71 / \log 2$

$= 6.71$ gears (rounded to 7 gears)

The arrangement of these 7 compounds gears is illustrated in Figure 3.16. These compound gears are connected by 6 idler gears in order to have a complete gear train structure. This idea is theoretically acceptable, but in practice, it is difficult to tell whether this idea is going to work or not.

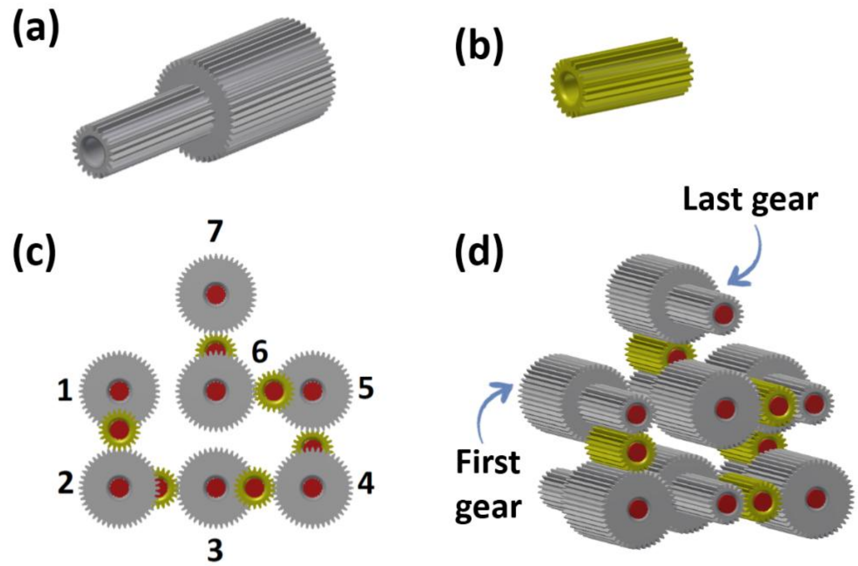


Figure 3.16: (a) The compound gear with ratio of 2, (b) The idler gear, (c) The gear train (2D view), and (d) The gear train (3D view)

3.6.3. Rough idea 3

The rough idea 3 focuses on weight concentration. The idea is to concentrate the weight as near as possible to the knee joint to reduce the required moment to flex the knee joint as illustrated in Figure 3.17. If the weight is concentrated at the inferior side (Figure 3.17 (a)), the distance between the knee joint and the center of concentrated weight is farther compared to the concentrated weight at the superior side (Figure 3.17 (b)). It is indirectly saying that the concentrated weight at the superior side is recommended in order to have a lower acted moment at the knee joint and to consume less energy to flex the knee. The actuated motor will be positioned upside down to allow more concentrated weight near to the knee joint as illustrated in Figure 3.18. A frame is needed to hold the motor in place.

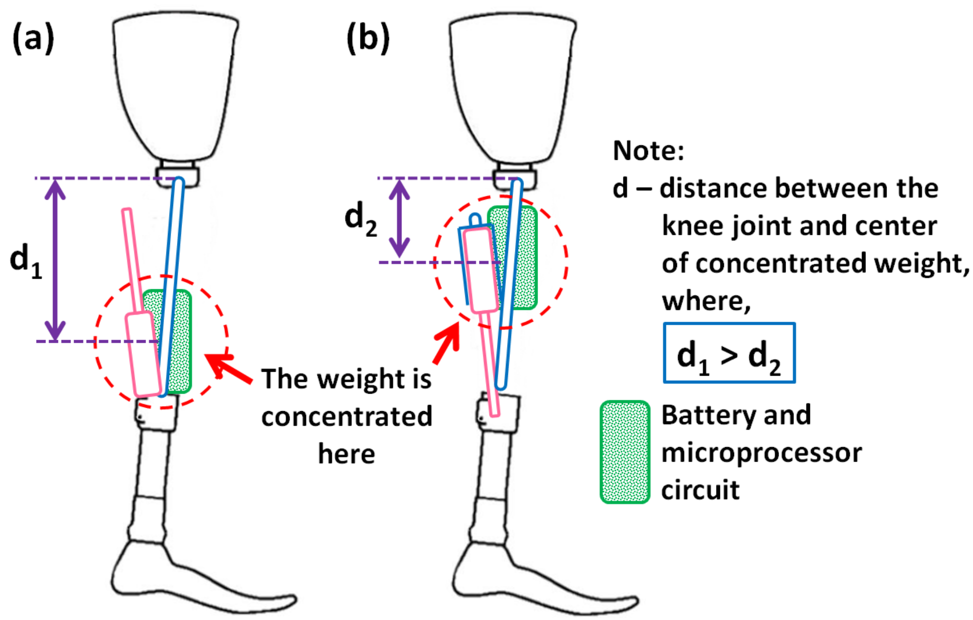


Figure 3.17: The weight concentration at (a) inferior side, and (b) superior side of the prosthetic knee

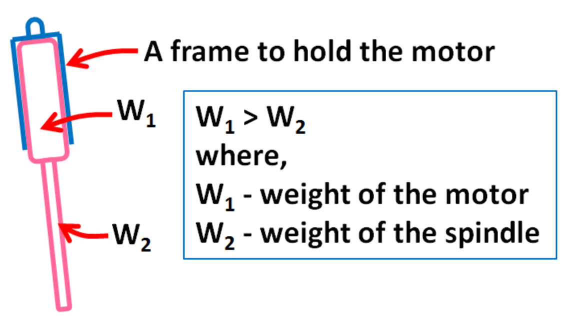


Figure 3.18 – The illustration of weight of the motor and the spindle

In order to allow the prosthetic knee to flex up to 120^{O}, a plastic chain will be used due its flexibility. The example of the plastic chain is shown in Figure 3.19. The third rough idea is illustrated in Figure 3.20. One end of the plastic chain is connected to the socket body, and the other end is connected to the motor frame. A pivot is positioned at the anterior side to create enough distance to create a moment at the knee joint.

Figure 3.19: Example of plastic chain (Reproduced from Tinycontrols, 2012)

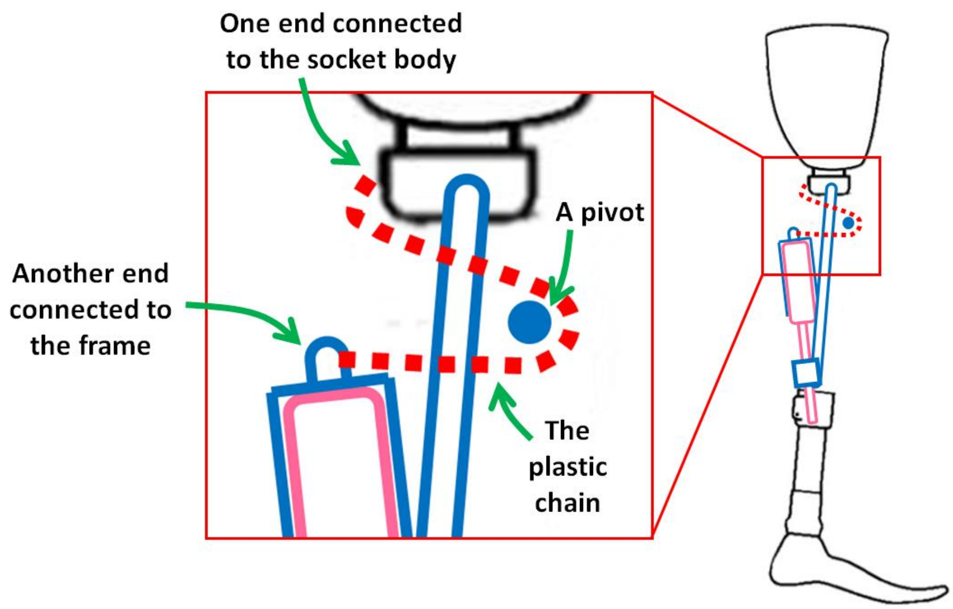


Figure 3.20: The configuration of the third rough idea

The illustration of the knee flexion at certain degrees is shown in Figure 3.21. As the weight is concentrated at the superior side, it is expected that the knee joint can be able to flex faster than the initial design.

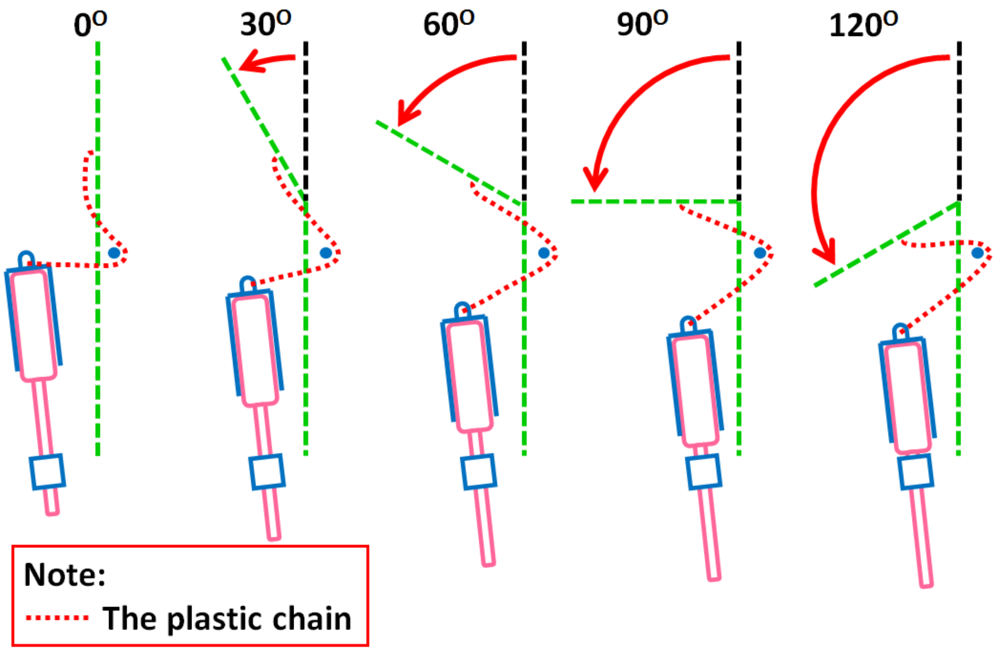


Figure 3.21: The illustration of knee flexion of the rough idea 3 at certain degrees

3.7. Finalize and Modelling of the Design

After the rough design stage was successfully went through, the proposed design must be finalized. The final design was obtained after considering advice and consultations from the supervisor and a few other mechanical engineers. A few discussions in terms of mechanical structure, strength, and ease of fabrication were carried out.

After the final design was finalized, the modelling of the design was drawn properly using a CAD software called SolidWorks 2013. This software is easy to use, and can be able to transfer the conceptual ideas of the prosthetic knee into 3D virtual prototype. This software works by creating individual parts of the prosthetic knee in separated files. After that, these individual parts were assembled in a separated file to have a complete virtual prototype of the prosthetic knee. The view of the SolidWorks 2013 software is shown in Figure 3.22.

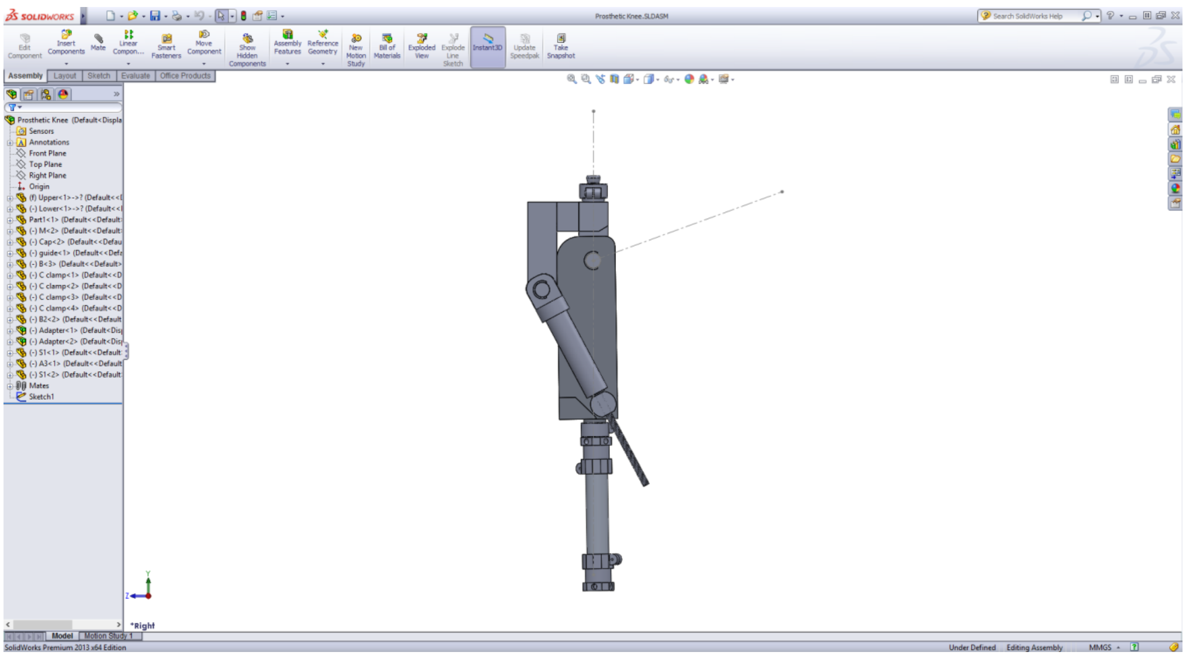


Figure 3.22: The view of SolidWorks 2013 software

3.8. *Simulation and Analysis of the Design*

After the final design was completely drawn, a simulation of the knee flexion from 0 to 120^{O} was performed. The same software of the SolidWorks 2013 was used. The software provides simulation tools to animate the movement of the prosthetic knee. It helps to test the design and to make any changes if required. The time spent for the prosthetic knee to flex was determined from the animation. The software provides a tool called 'Interference detection'. The tool identified any interferences between components of the prosthetic knee. It helped to evaluate and examine those interferences. By detecting these interferences, the final design was slightly changed (in terms of dimension and shape) in order to allow smooth movement of the prosthetic knee.

Another simulation was performed using Finite Element Analysis (FEA). The FEA is a simulation in terms of mathematical modelling by creating a situation to predict the outcome. In this project, the simulation was performed by creating a situation where an amputee with a weight of 80 Kg uses the prosthetic knee while lifting up his normal leg as illustrated in Figure 3.23. The selected situation is where the prosthetic knee experiences the highest load of body weight during walking.

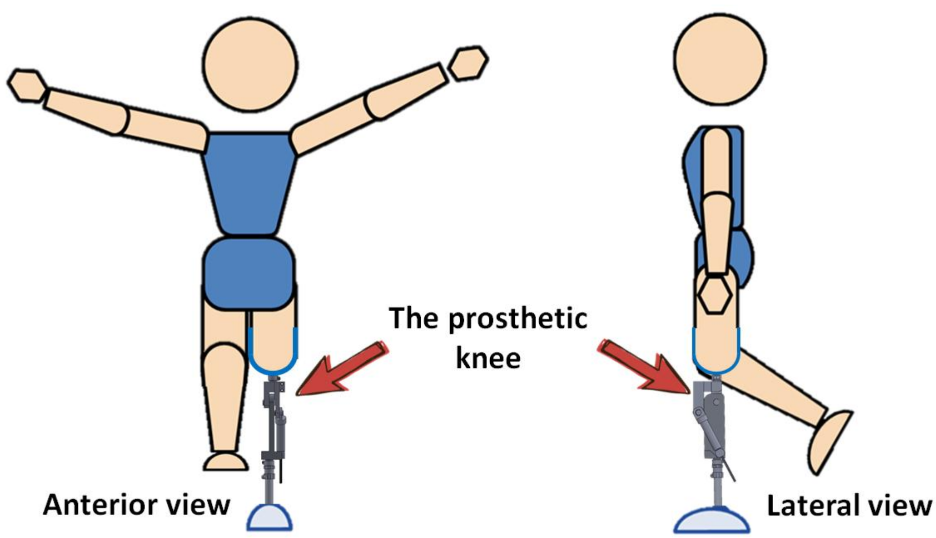


Figure 3.23: The situation of an amputee uses the prosthetic knee while lifting up his normal leg

The FEA was performed only on the prosthetic knee unit. A load of 800 N was applied at the most superior of the prosthetic knee unit as illustrated in Figure 3.24. Meanwhile, the most inferior part was set to be fixed. The results of the FEA were presented based on these two analyses. The analyses are displacements and stresses. The results were evaluated carefully.

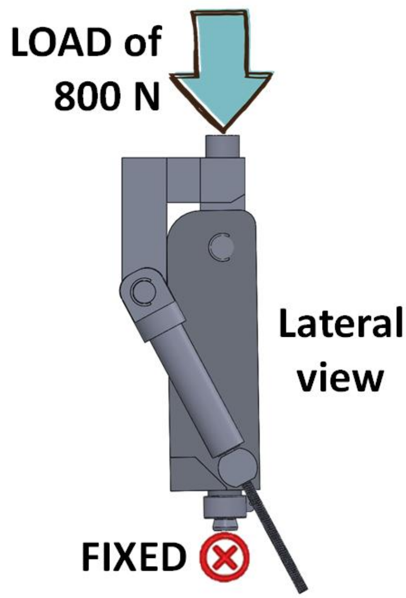


Figure 3.24: The application of a load of 800 N on the prosthetic knee unit

Chapter 4. RESULTS

4.1. *The Design*

4.1.1. Overview

The complete structure of the prosthetic leg of this project is illustrated in Figure 4.1. The socket, the pylon, and the prosthetic foot are commercial products from the market. This project focuses only in designing the prosthetic knee unit for transfemoral amputees. The prosthetic knee covers the areas of the thigh, and the shank. Figure 4.2 illustrates that 29.7% of total length of the prosthetic knee falls in the thigh area, and another 70.3% falls in the shank area. It means that the knee joint is located approximately 30% from the topmost of the prosthetic knee.

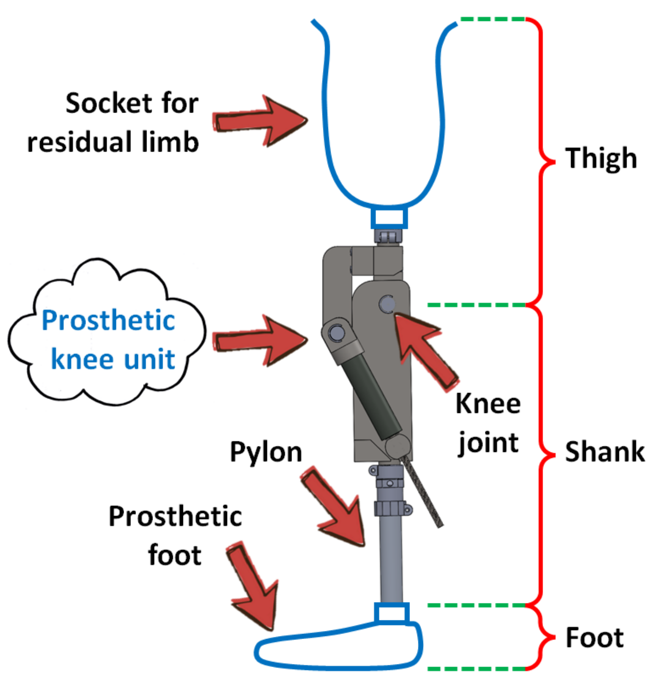


Figure 4.1: The complete structure of the prosthetic leg for this project

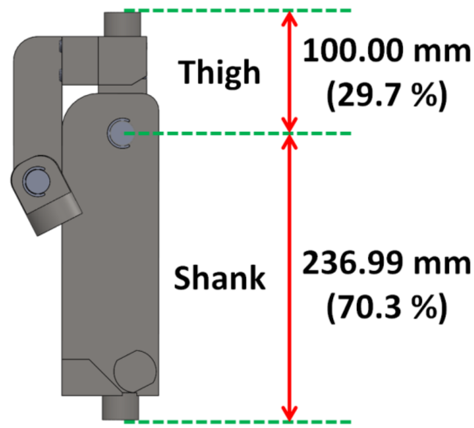


Figure 4.2: The percentage of the prosthetic knee at thigh and shank

Figure 4.3 shows the assembled and the exploded view of the prosthetic knee. All components of the prosthetic knee are labelled and shown in the figure. Each component has their own functions and characteristics.

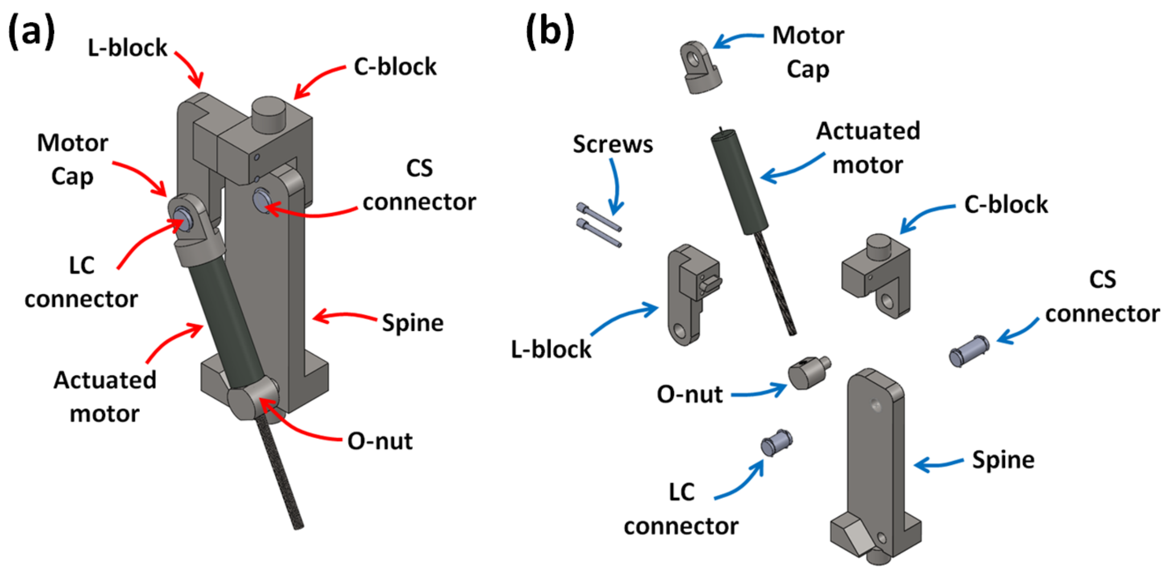


Figure 4.3: The assembled view (a) and the exploded view (b) of the prosthetic knee unit

4.1.2. The Blocks

There are two blocks in the design, the C-block and the L-block. Their name came from their shapes as illustrated in Figure 4.4. Meanwhile, Figure 4.5 illustrates how the blocks play a role as movable parts of the prosthetic knee to create moments at the knee joint. The actuated motor pushes the blocks and creates moments. These moments allow the prosthetic knee to flex.

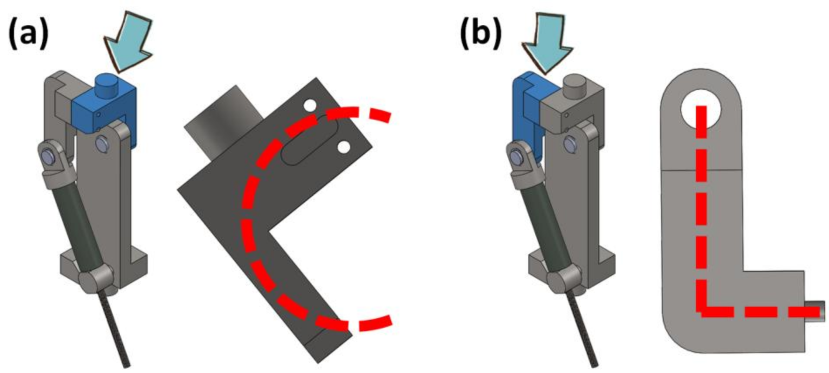


Figure 4.4: The shapes of the (a) C-block and (b) L-block

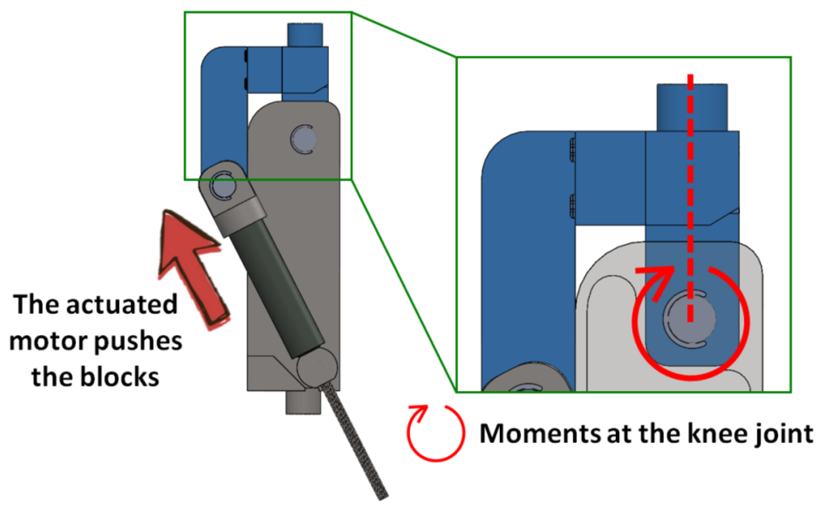


Figure 4.5: How the blocks create moments at the knee joint

These two blocks are attached by two screws as illustrated in Figure 4.6. The screws hold these two blocks together where they can be considered as one rigid body. They will be built separately for ease of fabrication. A male pyramid adapter (commercial product) is placed on top of the C-block. The male pyramid adapter will be connected to the socket. The CS connector connects the C-block to the spine. The L-block is connected to the motor cap using the LC connector.

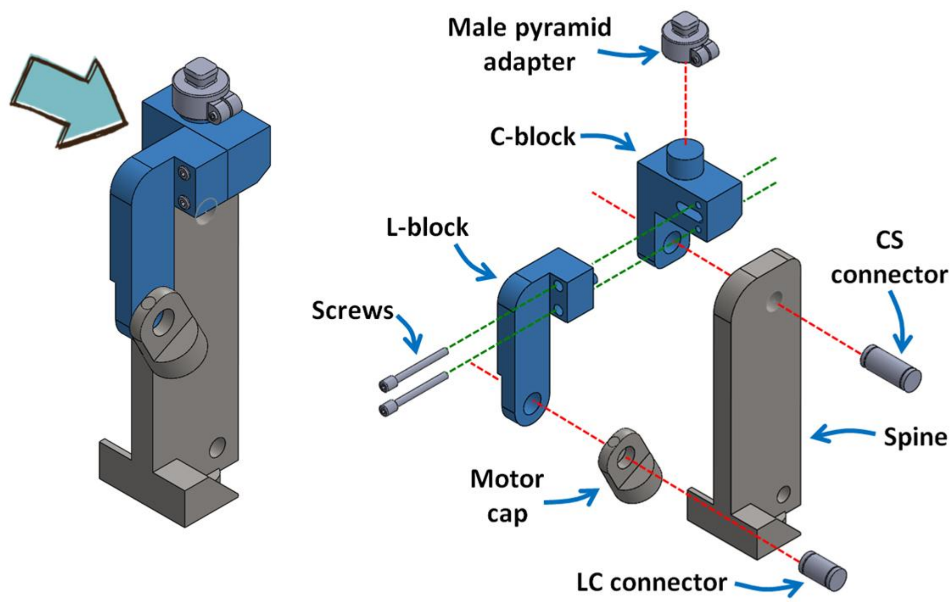


Figure 4.6: The connections of the blocks

4.1.3. The Spine

The spine acts as a backbone of the prosthetic knee where it supports most of the applied load on the prosthetic knee during any kind of activities. The spine is the biggest component in the prosthetic knee. There are two sides of the spine as illustrated in Figure

4.7. One side is the placement of the actuated motor. Meanwhile, the other side is where a battery and a microcontroller board are located. The microcontroller controls the movement of the actuated motor, and the battery supplies energy to the system. The connections of the spine are illustrated in Figure 4.8.

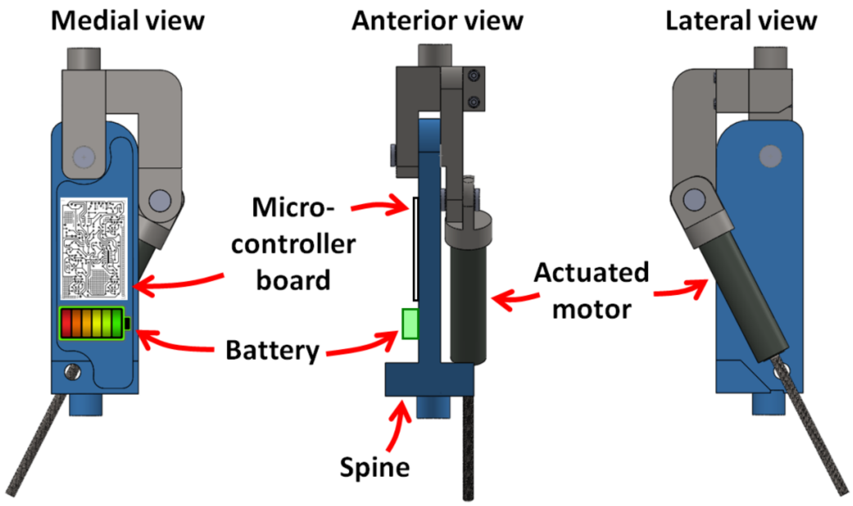


Figure 4.7: The two sides of the spine

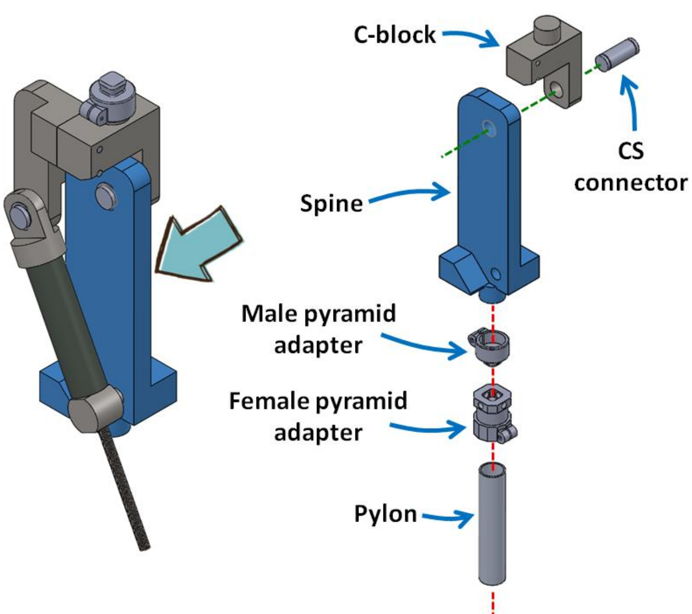


Figure 4.8: The connections of the spine

4.1.4. The Motor Cap and the O-nut

The motor cap and the O-nut hold the actuated motor in place. They work together to rotate the knee joint as illustrated in Figure 4.9. First, the microcontroller sends a signal to activate the actuated motor. Then, once the actuated motor is activated, the spindle rotates and creates a gap between the motor and the O-nut. The gap allows the motor to move upward. After that, the motor cap, which connected to the actuated motor pushes the blocks (both C-block and L-block) posteriorly. Finally, the knee joint rotates and allow the prosthetic knee to flex. The flexion of the prosthetic knee for certain degrees is illustrated in Figure 4.10.

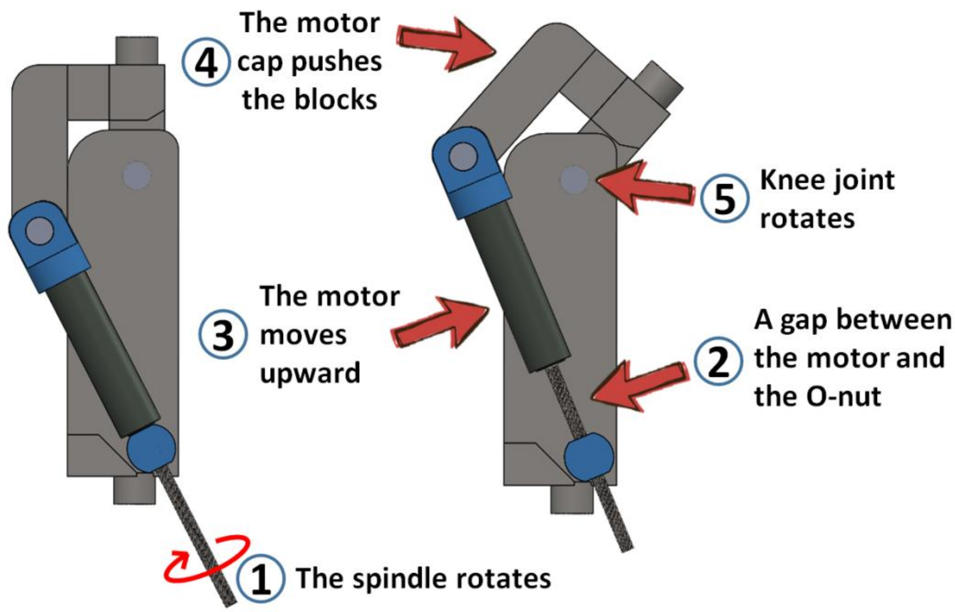


Figure 4.9: The role of the motor cap and the O-nut in rotating the knee joint

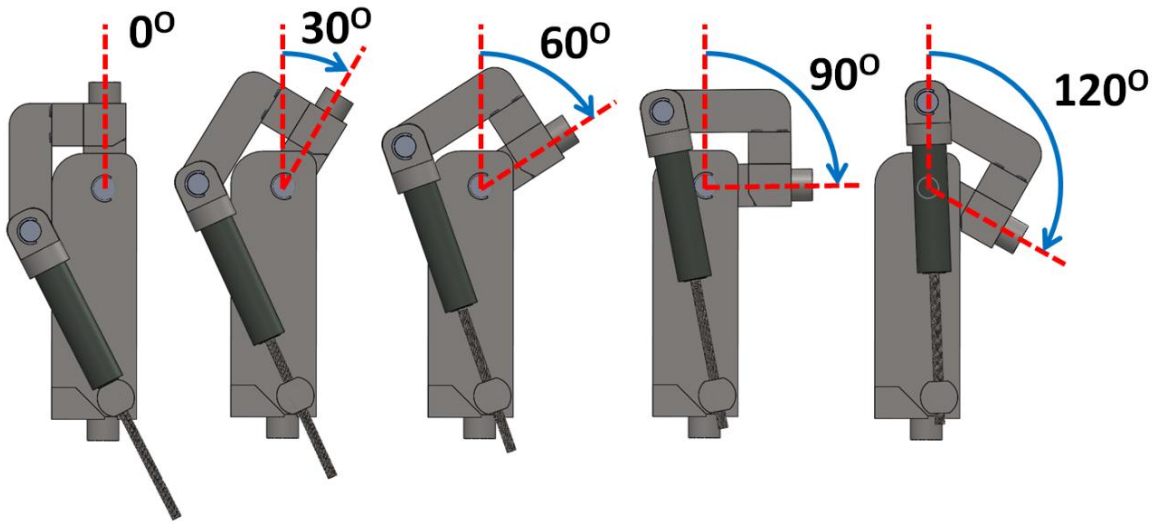


Figure 4.10: The flexion of the prosthetic knee at certain degrees

4.1.5. The Connectors

There are two main connectors in the design, the CS connector and the LC connector. The CS connector connects the C-block to the spine. Meanwhile, the LC connector connects the L-block to the motor cap. Two C-clamps are used to clamp each connector on both sides. Figure 4.11 illustrates how the CS connector is clamped by the C-clamps. Ease of maintenance can be achieved by using this type of clamping where the design can be easily assembled and disassembled. A lubricant or oil must be applied to the connectors to allow the prosthetic knee to move smoothly. This can be done by taking out the C-clamp and disassemble the components. The lubricant will be applied once for 3-4 months depending on how frequent the prosthetic knee is used.

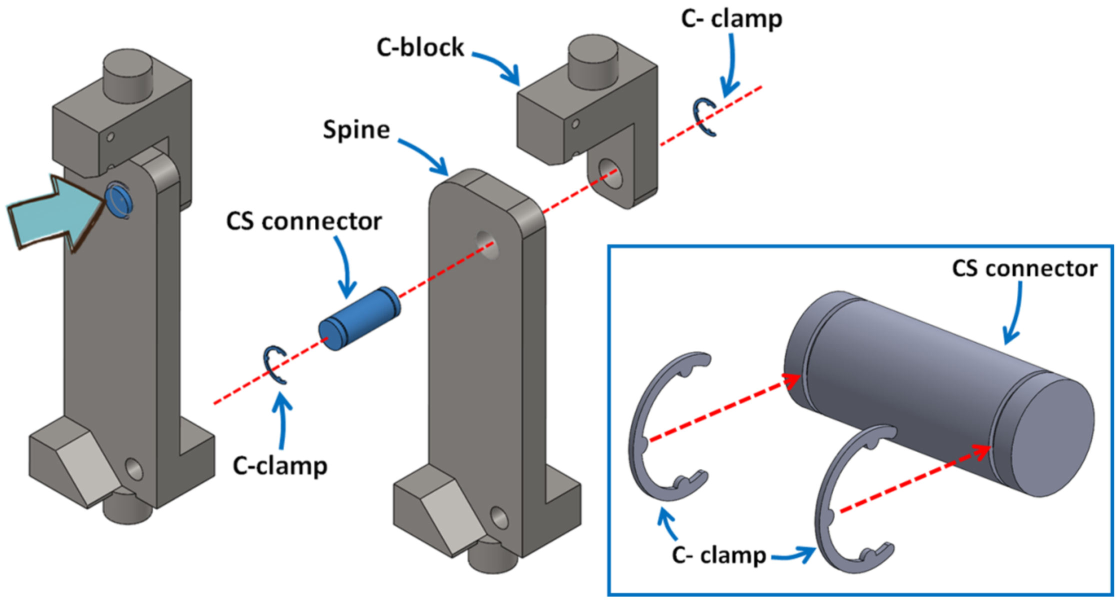


Figure 4.11: The orientation of the CS connector and the C-clamps

4.1.6. The Shank Cover

The shank cover was designed mainly for a cosmetic purpose. The shape of the shank cover is inspired by the shape of a normal shank as shown in Figure 4.12. It allows the prosthetic knee to look more appealing to the eyes. It also acts as a protection of the prosthetic knee from dust, dirt and damage. The shank cover consists of two symmetrical parts (A and B) as shown in Figure 4.13. The both parts are tightened by screws. Instead of old moulding method, the shank cover will be built by using 3D printing technology. A 3D printer can print out the shank cover in different sizes (depends on the size of amputees), and colors (depends on the amputees' preferences). The material of the shank cover is a type of plastic known as acrylonitrile butadiene styrene (ABS). The ABS was picked because it is a dominant material in the 3D printing technology.

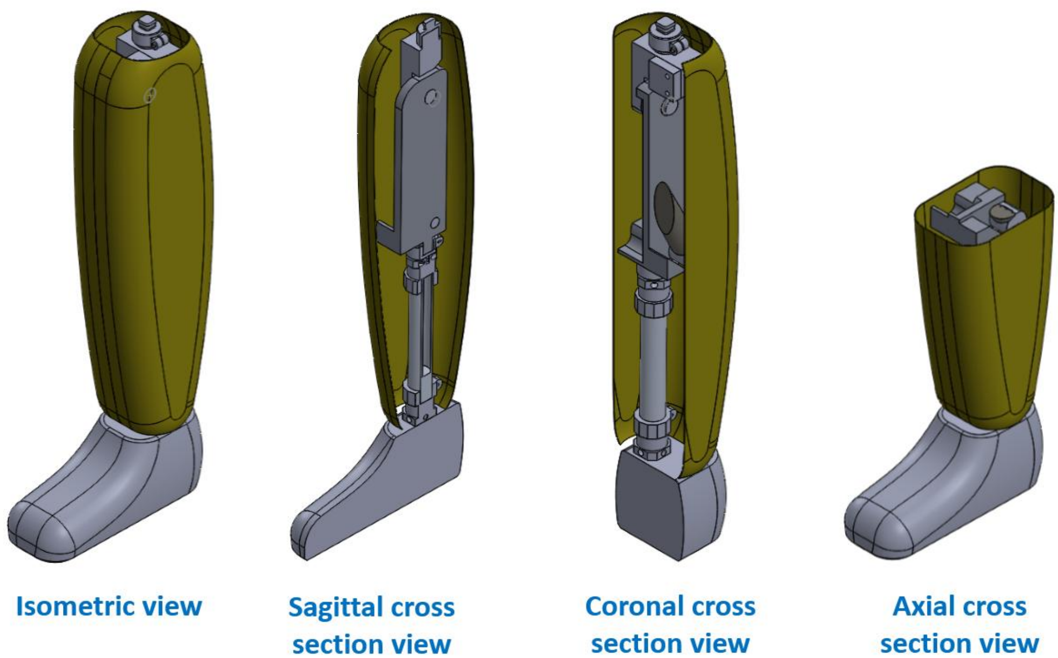


Figure 4.12: The different views of the shank cover

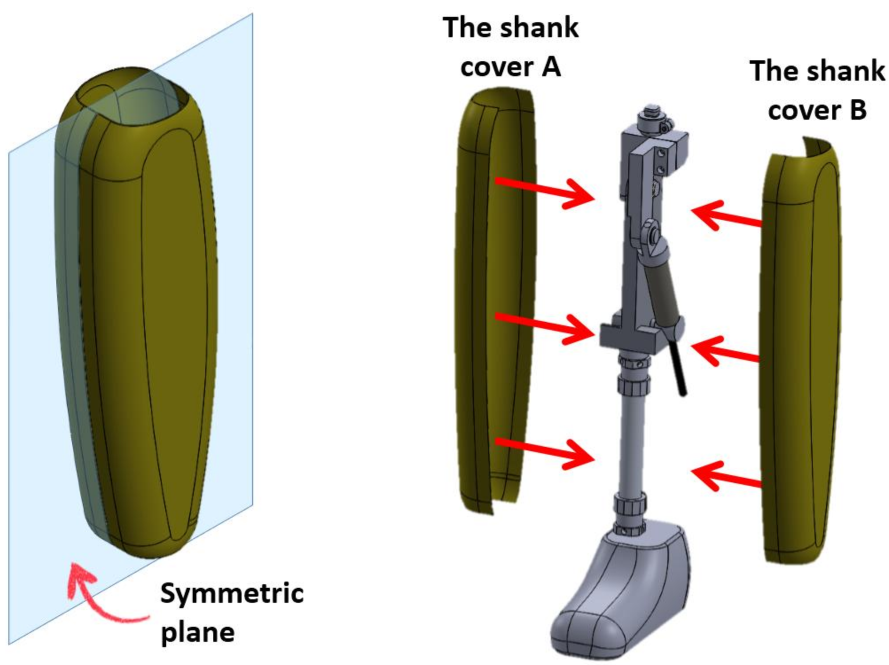


Figure 4.13: The symmetrical parts of the shank cover

4.2. The Simulation and Analysis

4.2.1. Finite Element Analysis

Three types of materials are nominated in this project. They are Aluminium Alloy 6061, AISI 316 Stainless Steel, and AISI 4130 Steel. In the end of this project, only one material was selected. The simulations using FEA were conducted to all materials. Two types of analyses were performed. They were the stress Von Mises analysis and the displacement analysis as shown in Figure 4.14 and Figure 4.15. The summary of the static stress analysis and the displacement analysis are tabulated in Table 4.1.

4.2.1.1. The Stress Von Mises Analysis

The stress distribution experienced by the prosthetic knee unit ranged from 4,246.9 N/m^2 to 30,177,486.0 N/m^2 for Aluminium Alloy 6061, 2,990.0 N/m^2 to 31,657,876.0 N/m^2 for AISI 316 Stainless Steel, and 2,857.1 N/m^2 to 31,865,922 N/m^2 for AISI 4130 Steel. The color key refers to the different stress value distributed across the prosthetic knee unit. Low stress distribution is highlighted in blue and light blue, which covered almost every part of the prosthetic knee unit. High stress distribution (highlighted in red and yellow) is present in the gap between the C-block and the spine as shown in Figure 4.16. Failure might happen to that part if excessive load is applied to the prosthetic knee unit. From the static stress analysis, the AISI 4130 is the best material to be selected because there is a big gap between the values of the maximum stress (31,865,922.0 N/m^2) to the value of yield strength (460,000,000.0 N/m^2) of that material. The yield strength is the value of stress of the material can withstand without any permanent deformation.

4.2.1.2. The Displacement Analysis

From Figure 4.15, the color key refers to the different displacement value distributed across the prosthetic knee unit. As the bottom of the prosthetic knee unit is set to be fixed, there is zero displacement on that region. The displacement is high in red/orange color regions which distributed at top regions for all types of materials. The highest displacements recorded are 0.319 mm for Aluminium Alloy 6016. Then it followed by 0.115 mm for AISI 316 Stainless Steel, and 0.109 mm for AISI 4130 Steel. The displacements for both AISI 316 Stainless Steel and AISI 4130 Steel are very small and preferable. Although the Aluminium Alloy 6016 recorded the highest displacement of 0.319 mm, it is still acceptable because the value is still very small and almost negligible.

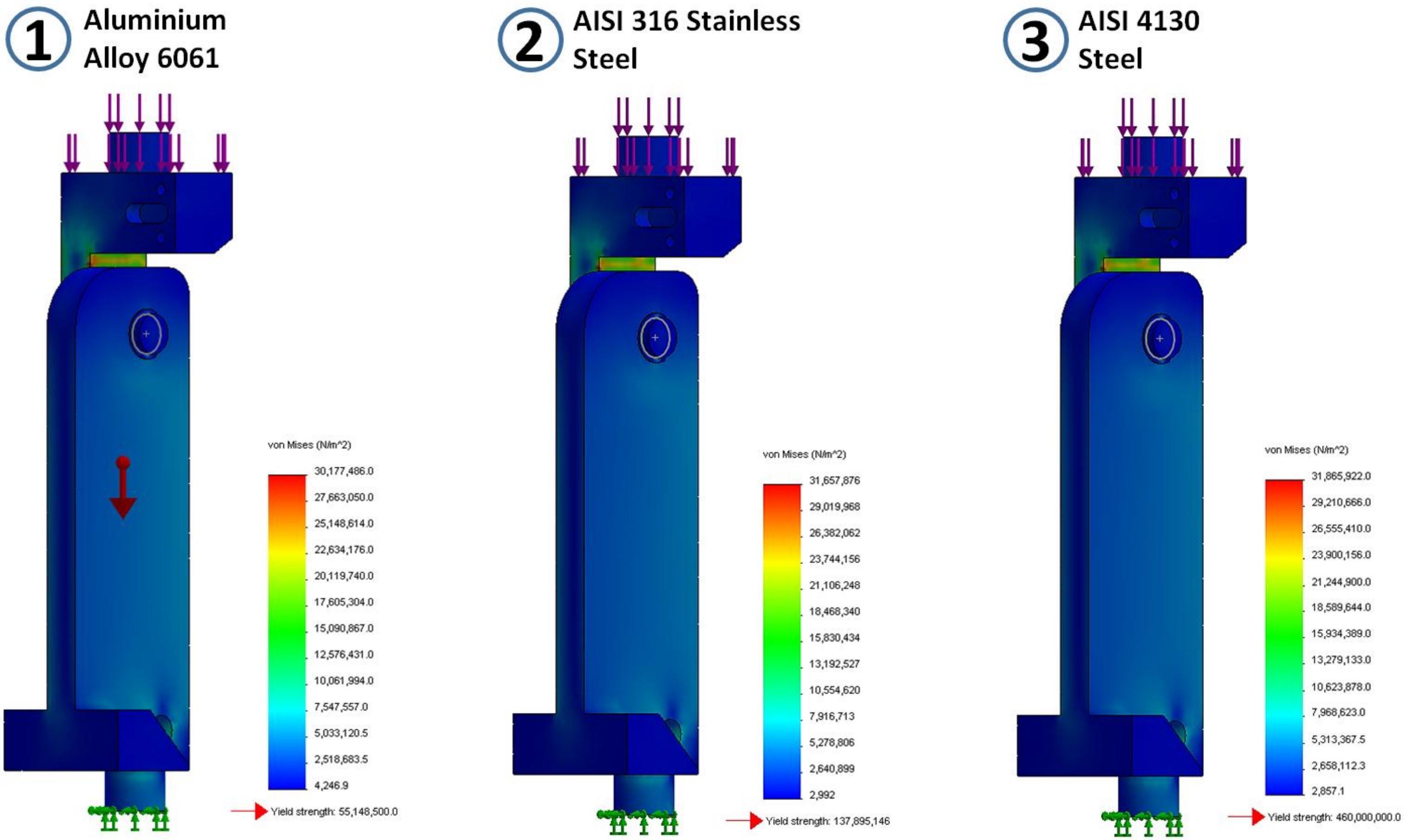


Figure 4.14: The stress Von Mises analysis for all three materials

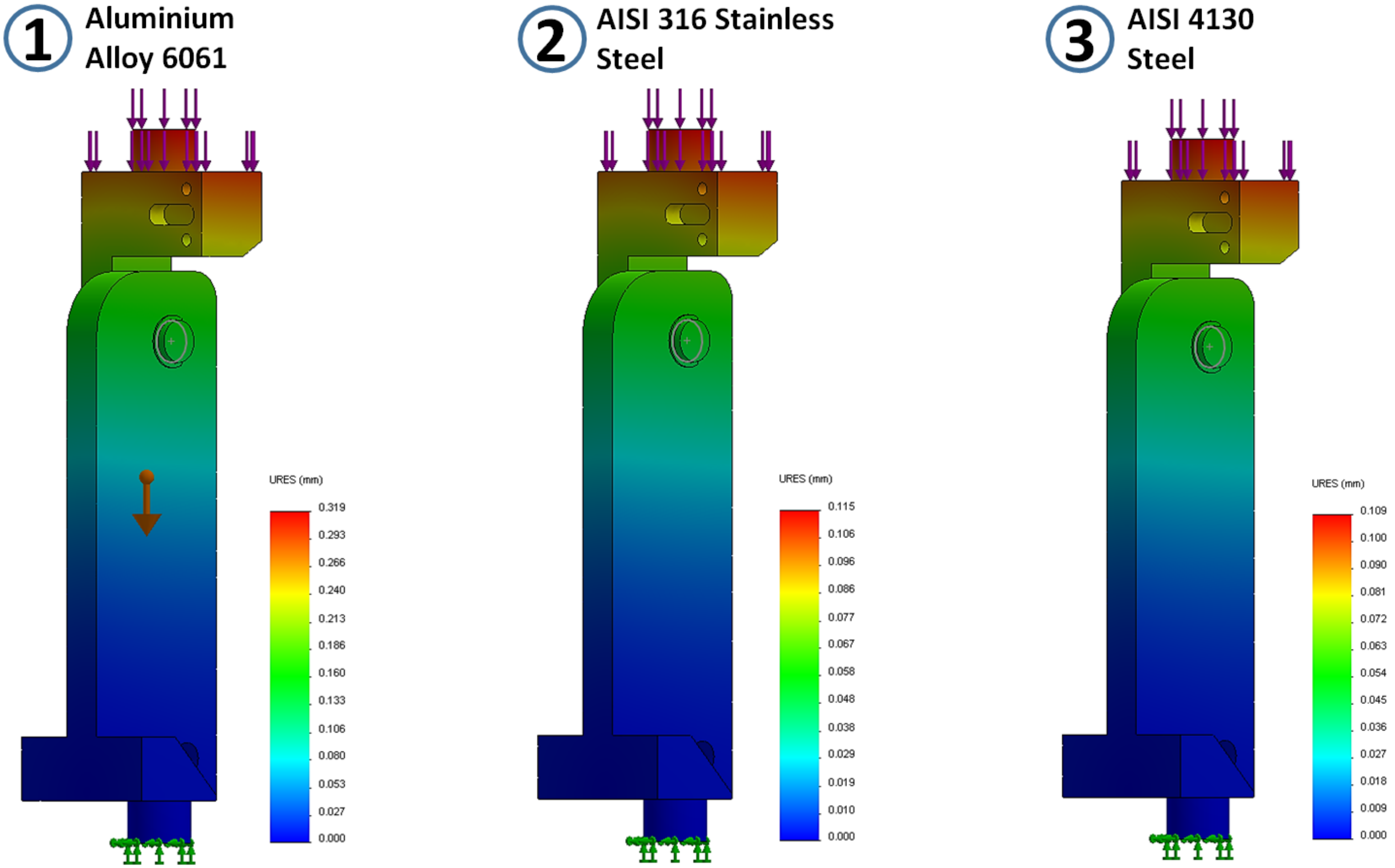


Figure 4.15: The displacement analysis of all three materials

Table 4.1 - The summary of the static stress analysis and the displacement analysis

Materials	Static stress analysis (N/m^2)			Displacement analysis (mm)	
	Min	Max	Yield strength	Min	Max
Aluminium Alloy 6061	4,246.9	30,177,486.0	55,148,500.0	0	0.319
AISI 316 Stainless Steel	2,990.0	31,657,876.0	137,895,146.0	0	0.115
AISI 4130 Steel	2,857.1	31,865,922.0	460,000,000.0	0	0.109

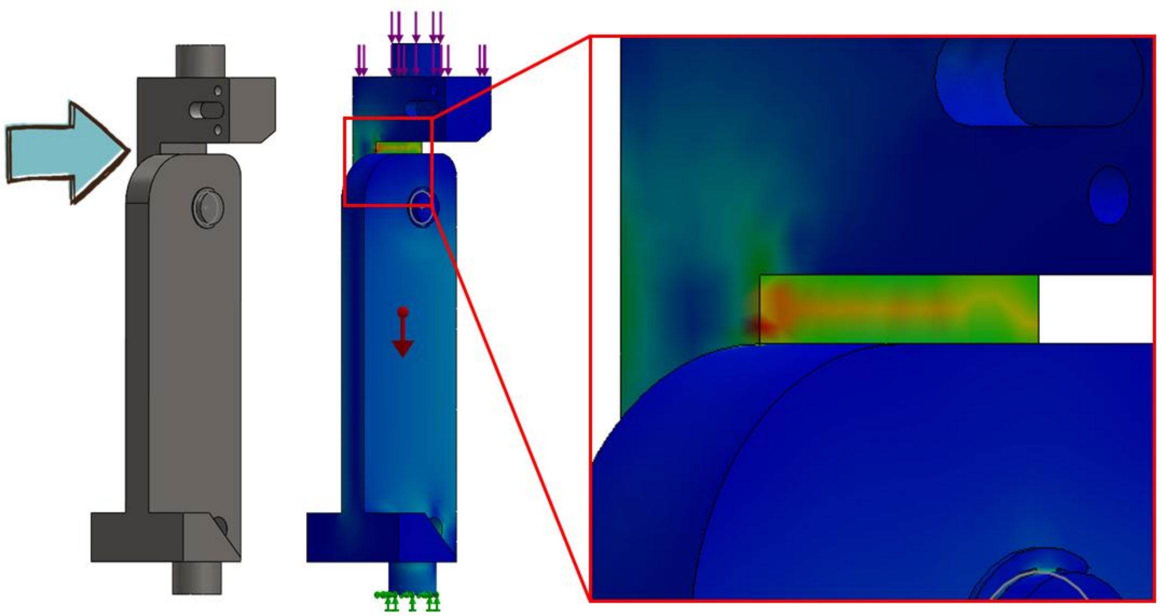


Figure 4.16: High stress distribution at the gap between the C-block and the spine

4.2.2. **Knee Flexion Animation**

The SolidWorks 2013 was used to animate the prosthetic knee in order to explain and simulate its function and assembly. The animation was converted to a video with format of Audio Video Interleave (AVI). The AVI is a popular container file format used for watching standard definition video and can be played in almost all video players. To obtain a high quality video, the resolution of the video was set to 1920x1080 with a frame rate of 7 frames/second. The video is used to present the idea of the design. It also can be used as a bridge to communicate between technical and nontechnical people. The video shows the animation of the movement of flexing the prosthetic knee from 0 – 120^{O}. The design was simulated by considering the specifications of the actuated motor. From the animation video, the prosthetic knee flexes from 0 – 120^{O} within 6 seconds. Some screenshots of the animation video are shown in Figure 4.17.

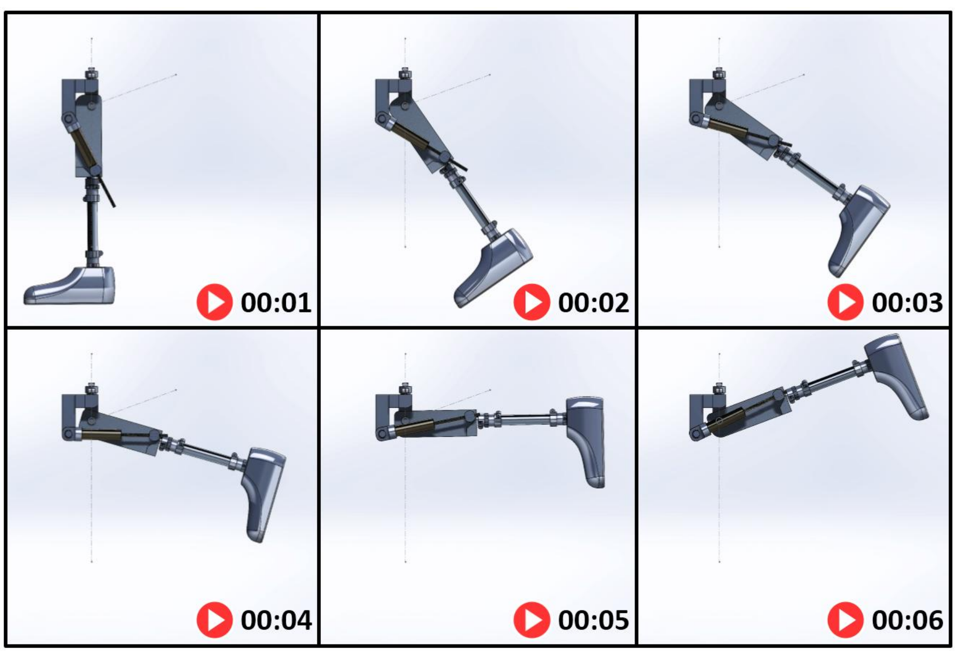


Figure 4.17: Some screenshots of the animation video

4.3. *Estimated Total Weight and Total Cost for Fabrication*

The estimated total weight and the estimated total cost of fabrication are tabulated in Table 4.2. The estimated weights of each part were calculated by the SolidWorks 2013. The estimated total weight of raw materials is the estimation of the total weight of the parts to be fabricated as shown in Figure 4.18. Meanwhile, the estimated total weight of the prosthetic knee is a summation of the estimated total weight of raw materials, the estimated weight of the shank cover, and the estimated total weight of the available parts as illustrated in Figure 4.19 and Figure 4.20.

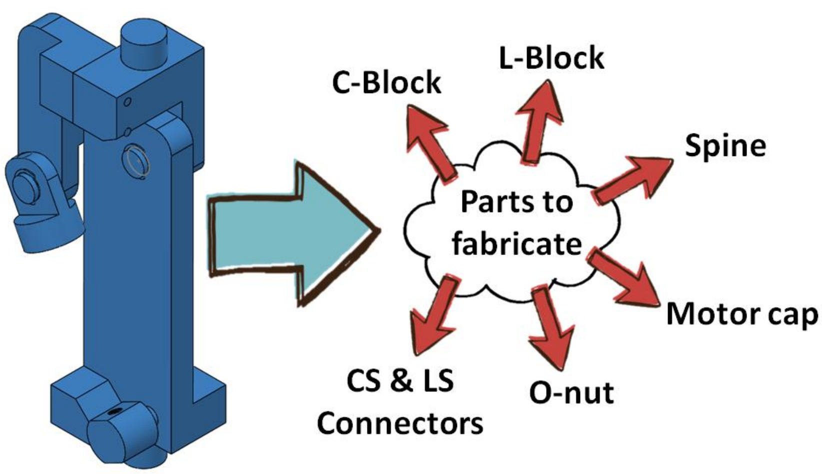


Figure 4.18: The parts to be fabricated

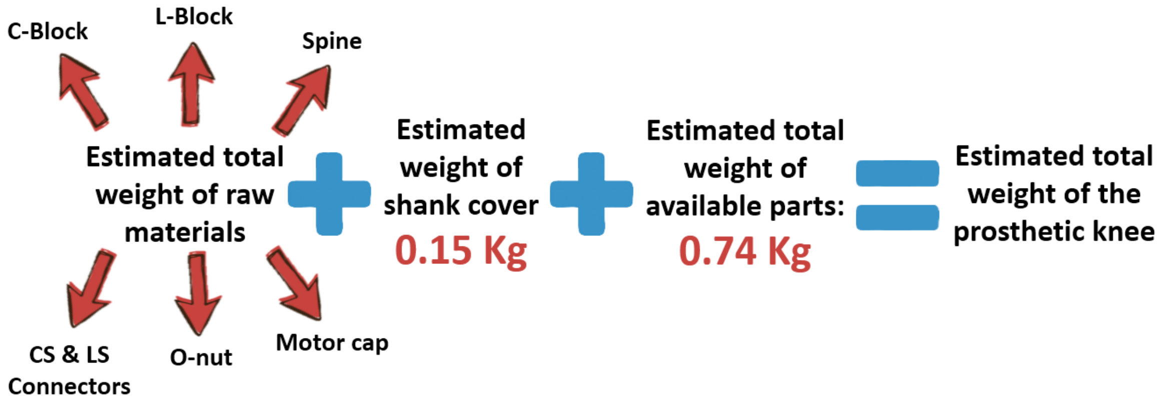


Figure 4.19: The formula in estimating the total weight of the prosthetic knee

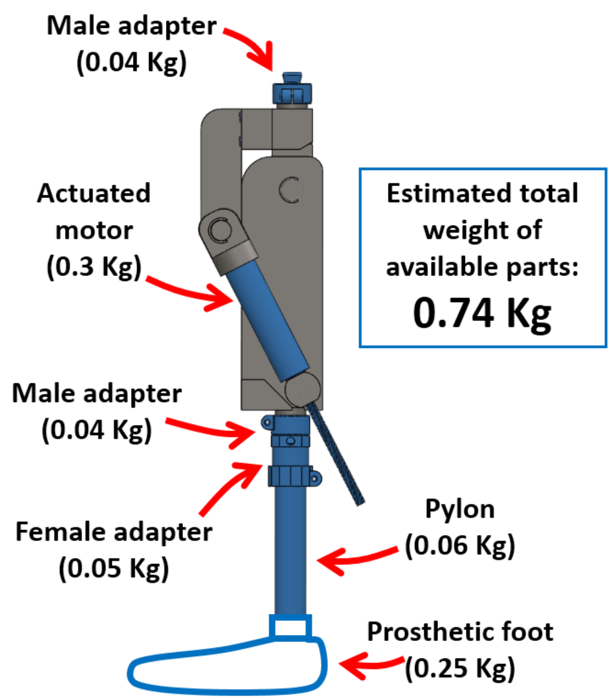


Figure 4.20: The estimation of weight of the available parts

The estimated total weight of raw materials are vary from Aluminium Alloy 6061 to AISI 4130 Steel. The Aluminium Alloy 6061 is the lightest material compared to the other two materials with weight of 2.02 Kg. The estimated weight of raw materials for both AISI 316 Stainless Steel and AISI 4130 Steel are more or less the same with weight of 5.99 Kg and 5.86 Kg respectively. The spine is the heaviest component among all components in the prosthetic knee. After assembling all components with available parts and shank cover, the lightest estimated weight of the prosthetic knee is using Aluminium Alloy 6061 with weight of 2.87 Kg. By comparing all three materials, the Aluminium Alloy 6016 is the only material that fulfil the fourth design criteria to have a weight of prosthetic knee less than 3 Kg.

The estimated total cost of raw materials were calculated based on the prices available in the SolidWorks 2013 database. The price of a Kg of each material were converted from US Dollar (USD) to Ringgit Malaysia (MYR). The cheapest estimated cost of raw material goes to the AISI 4130 Steel with a cost of RM 9.84. The fabrication charges of RM 1800 is the estimated cost from Zecttron Sdn Bhd. It includes the fabrication of all components in the prosthetic knee. After considering all the simulation results, the Aluminium Alloy 6061 was selected to be used in the design. Although the Aluminium Alloy 6061 fell behind for the FEA results, its lightweight and cost efficiency made it to be the best material for the prosthetic knee. The shank cover will be built separately by using a 3D printer. The 3D printed ABS costs RM2 per gram (3DCipta, 2013).

Table 4.2: The estimated weight and cost for different types of materials

Materials	Aluminium Alloy 6061	AISI 316 Stainless Steel	AISI 4130 Steel
Parts	The estimated weight of each component (Kg)		
C-block	0.47	1.40	1.37
L-block	0.38	1.13	1.11
Spine	0.93	2.76	2.70
Motor cap	0.09	0.26	0.25
O-nut	0.08	0.23	0.22
CS connector	0.04	0.12	0.12
LC connector	0.03	0.09	0.09
Estimated total weight of raw materials	2.02	5.99	5.86
Estimated total weight of the prosthetic knee	2.87	6.84	6.71
	Price of raw materials (RM/Kg)		
	7.26	11.81	1.68
	The estimated cost (RM)		
Estimated total cost of raw materials	14.67	70.74	9.84
Estimated total cost of fabrication	2114.67	2170.74	2109.84

Notes:

- The total cost of fabrication including the estimated fabrication charges of RM 1800 and the cost of the shank cover (RM 2 per gram X 0.15 Kg = RM 300)
- The price of the shank cover was referred to the price provided by 3DCipta (2013).

Chapter 5. DISCUSSIONS

5.1. The Design

5.1.1. Overview

The final design was formulated after taking advice from the supervisor and consultation with a few other mechanical engineers. All three rough ideas were taken into account during the design process. The comparisons of all three rough ideas are summarized in Table 5.1. The gearing structure of rough idea 1 and rough idea 2 were rejected due to their complex structures. The complexity causes high cost of fabrication and high maintenance. The idea to change the pitch to pitch value was rejected too. The spindle comes together with the actuated motor. Changing the spindle may change the overall performance of the actuated motor. The chain structure was also rejected due to high maintenance. In the end, the idea of placing the actuated motor upside down was accepted. But, the placement of the actuated motor was changed from posterior (rough idea) to anterior side. By placing the actuated motor anteriorly, the prosthetic knee has more freedom to flex up to 120^{O}.

Table 5.1: The comparison of three rough ideas

	Rough idea 1	Rough idea 2	Rough idea 3
Gearing structure	Modified internal gearing structure ✗	Complex multiple compound gears and idler gears ✗	

Note: ✓ Accepted, ✗ Rejected

Table 5.1, continued

	Rough idea 1	Rough idea 2	Rough idea 3
Change the pitch to pitch value of the spindle	From 1 mm to 2 mm ✖		
Inverted placement of the actuated motor			The actuated motor is place upside down ✔
Chain structure			Using plastic chain ✖

Note: ✔ Accepted, ✖ Rejected

5.1.2. The Design Criteria

In general, the final design has fulfilled almost all the design criteria as shown in Table 5.2. The first design criteria was achieved where the design has a mechanism that allowed the prosthetic knee to flex from 0 – 120^{O}. From the animation, the time taken for the prosthetic knee to flex from 0 – 120^{O} is within 6 seconds. This value does not fulfil the second design criteria, but it is close enough. In terms of dimension, the length from the knee joint to the most bottom of the prosthetic knee is 236.99 mm as illustrated in Figure 5.1. The width of the prosthetic knee is around 75 mm instead of 70 mm from the design criteria. Although there is a difference in the width value, 75 mm is still acceptable and consider as fulfilling the third design criteria. Since the length of the prosthetic knee is less than 250 mm, it is expected that Malaysian amputees would be able to use the prosthetic knee.

The Aluminium Alloy 6016 was selected as the material to be used to fabricate the prosthetic knee. The estimated total weight of the prosthetic knee is 2.87 Kg. This fulfil the fourth design criteria where the design should have a weight of less than 3 Kg. From the FEA results, the prosthetic knee was proven to be able to withstand a load of 800 N. It achieved the goal of fifth design criteria. It is expected that an amputee weighted less than 80 Kg could be able to use the prosthetic knee. In term of cost, the total estimated cost to fabricate the prosthetic knee is RM 2114.67. By selecting the Aluminium Alloy 6061, this value is still affordable while maintaining the overall performance of the prosthetic knee. The appearance of the prosthetic knee looks neat and appealing by covering the prosthetic knee unit with the shank cover. With the affordable cost and the aesthetic value, the sixth design criteria was achieved too.

Table 5.2: The design criteria checklist

DC 1: Range of motion	
The prosthetic knee must be able to flex from 0 to 120 degrees.	✓
DC 2: Speed	
The time taken to flex the prosthetic knee from 0 to 120 degrees must be within 2 seconds.	✗
DC 3: Dimension	
The prosthetic knee must be small enough to be used by Malaysian amputees.	✓
The length and the width of the prosthetic knee should be around 250 mm and 70 mm respectively.	✓
DC 4: Weight	
The prosthetic knee must be light. The materials used should be picked carefully to avoid excessive weight.	✓
The prosthetic knee should have a weight of maximum 3 Kg.	✓

Table 5.2, continued

DC 5: Strength	
The prosthetic knee should be able to withstand a load of 800 N.	✓
The amputees weighted of 80 Kg and less should be able to wear the prosthetic knee.	✓
DC 6: Cost and quality	
The materials used must be selected based on their cost and quality.	✓
The aesthetic value must be taken into account during the design process.	✓

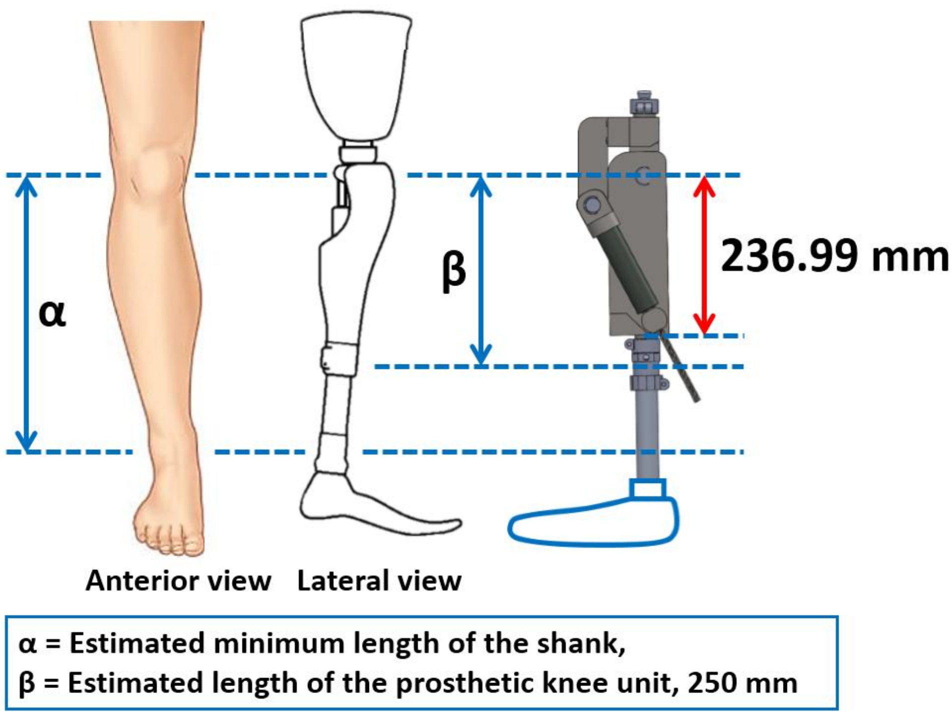


Figure 5.1: The length from the knee joint to the bottom of the prosthetic knee

5.2. The Simulation

The simulation process was completed by using the SolidWorks 2013. The simulation was required to ensure that the product would not fail under real life situation. The simulation of FEA was performed to both the C-block and the spine only. It is because only these two components are aligned with the direction of load from the amputee's body weight as illustrated in Figure 5.2. The rest of the components are located at the periphery of the prosthetic knee which is not affected by the load. From the FEA results, it shows that the prosthetic knee is safe to be used for amputees of weight less than 80 Kg. There is an area which might fail if excessive load is applied onto it. It is located at the gap between the C-block and the spine.

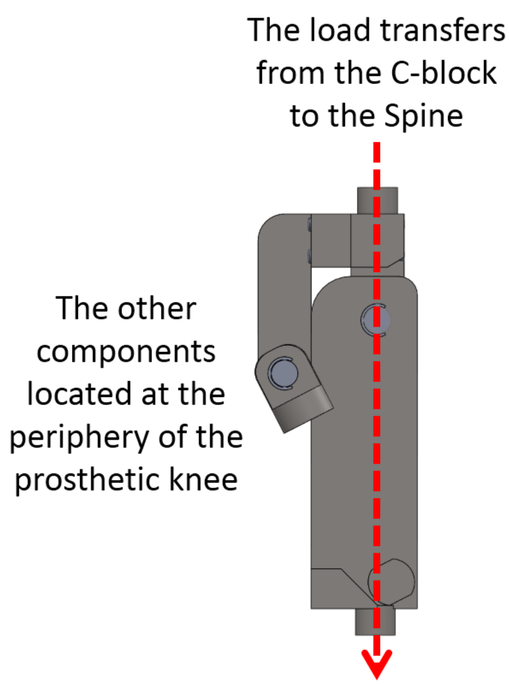


Figure 5.2: The load experience by the C-block and the spine

5.3. Suggestion for Future Research

Since the second design criteria was not achieved, more work needs to be done to fulfil the goal of having the knee flexes within 2 seconds from 0 – 120^{O}. To increase the speed while using the available actuated motor, a gearing structure with a simple design is needed. The rough ideas of this project introduced complex gearing structures where they are difficult to maintain and high cost to fabricate. For the future research, a simple gearing structure that allows low maintenance and low cost to fabricate should be implemented into the prosthetic knee. Instead of gearing structure, there are a few more things that can be implemented such as the use of chains, cable, and belts.

The specifications of the actuated motor need to be revised. It is pretty obvious that the available actuated motor has a very high torque, but low in speed. The actuated motor was purchased with focusing too much on the torque and overlooked at the speed. For the future research, the new specifications of the actuated motor should have a balance between the torque and the speed. This can be done by purchasing a new actuated motor or modifying the available motor until it achieves the revised specifications.

Chapter 6 Conclusion

Prosthetic knee design involves making a replacement of a missing lower limb by mimicking a normal, healthy knee. The prosthetic knee should be functional, comfortable, and good in visual appearance. This project is just a small portion of a big study of a microcontroller prosthetic leg. The study has its own initial design of the prosthetic knee. There are a few downsides of the initial design of the prosthetic knee that need to be improved. The major downsides are low in speed to flex the knee, low in aesthetic value, and bulky. This project targeted to overcome those problems. The design criteria were listed down as guidelines during the design process. All design criteria were achieved except one. Three different types of materials were simulated by using FEA technique. In the end, the Aluminium Alloy 6061 was selected to be the best material (among the three types of materials) to be used to fabricate the prosthetic knee. It could withstand a load of 800 N without fail. The maximum displacement of the prosthetic knee after applying 800 N is 0.319 mm. This value is very small and can be ignored. The estimated total weight of the prosthetic knee is 2.87 Kg. The shank cover made the prosthetic knee to look more appealing. The fastest time for the proposed design to flex from 0 to 120^{O} was 6 seconds. The quality of the prosthetic knees in the market is much higher compared with the proposed prosthetic knee from this project. But, in terms of functionality, both of them are still similar. The estimated total cost of fabrication of the prosthetic knee is RM 2114.67. This amount is affordable to be bought by amputees who come from low to medium income family.

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APPENDIX A1 – THE MAXON SPINDLE DRIVE CATALOG (FRONT PAGE)

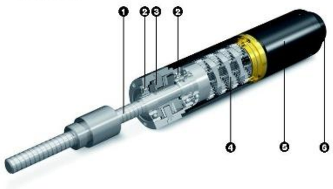
APPENDIX A2 – THE SPINDLE DRIVE CATALOG (BASICS)

Spindle Drive Basics

NEW

Design

1. Spindle, directly implemented in the gearhead
2. Radial bearing
3. Axial bearing
4. Planetary gearhead 0 - 4 stages
5. Motor
6. Encoder



The particular type of spindle required must first be established before a spindle drive can be designed. Every type of spindle has different characteristics and a number of specific limits. These limits are taken into account in the technical data.

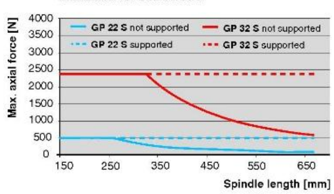
Ball screw:
- highly efficient
- not self-locking
- high load capacity

Metric spindle:
- self-locking
- low costs

Trapezoidal spindle:
- same as metric spindle
- higher load capacity than metric spindle

Feed force
For the calculation of the feed force acceleration and friction forces as well as gravity have to be taken into consideration. Exceeding the maximum permissible load must be avoided, as this damages the spindle. The maximum permissible feed force is displayed for standard spindles. For longer spindles, the permissible feed force can be limited by the critical compressive force of the spindle. In this case, supporting the end of the spindle may be necessary.

Limitation for ball screws



Torque
The required torque of the spindle M_a [mNm] is calculated with the feed force F_a [N] (load), the thread pitch p [mm] and the efficiency of the spindle η_1.

$$M_a = \frac{F_a \cdot p}{2 \cdot \pi \cdot \eta_1}$$

In combination with the gearhead, the required motor torque M_{mot} [mNm] is:

$$M_{mot} = \frac{F_a \cdot p}{2 \cdot \pi \cdot i \cdot \eta}$$

Where i is the gearhead reduction ratio and η the efficiency of the complete spindle drive.

Technical Data

The "Technical Data" block contains generally applicable data on spindle, nut and gearhead. These are independent of the gearhead reduction ratio.

Length
The data sheets show the spindle drives with the standard lengths. Other lengths are available as an option in 5 mm steps up to a given maximum length. Please give detailed requirements for special lengths.

Max. efficiency / mass inertia
The values stated refer to the spindle alone (without gearhead). The values with gearhead are given in the „Gearhead data" main data field.

Nut
Standard spindle drives are supplied with a thread nut. Flange or cylinder nuts are also available as an option. See details with corresponding reference number on page 8.

Bearing
The output stage and the spindle are supported by preloaded axial bearings. This means that the high axial forces can be absorbed directly by the gearhead without additional support.

Speed and feed velocity
Feed velocity v [m/s] is linked to output speed n [rpm] by the pitch p [mm].

$$v = \frac{p \cdot n}{60}$$

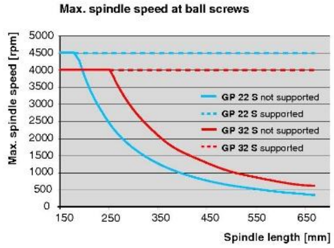
In combination with the gearhead, the motor speed n_{mot} [rpm] is:

$$n_{mot} = \frac{v \cdot 60 \cdot i}{p}$$

Where i is the gearhead reduction ratio and p the spindle pitch.

The spindle speed is limited by the resonance frequency of the spindle and for ball screws additionally by the ball return system.

Max. spindle speed at ball screws



In addition, the maximum permissible speed of the gearhead has to be considered.

Spindle Drive Data

Reduction ratio
The reduction ratio indicates the ratio by which the speed of the spindle is lower than the motor speed.

Absolute reduction ratio
Provides the reduction ratio as an exact ratio of two integers.

Max. feed force (continuous)
Is the maximum permissible feed force which may be continuously applied. Exceeding this value results in a reduced service life.

Max. feed force (intermittent)
Is the maximum permissible feed force which may be intermittently applied. „Intermittently" is defined as follows:
- during max. 1 second
- during max. 10% of operation

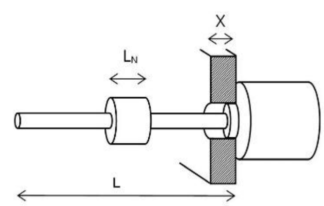
Exceeding these values results in a reduced service life.

Max. efficiency
The given efficiency is a maximum value that applies when loaded with maximum feed force. Efficiency falls sharply with very small loads. The stated value refers to the complete spindle drive (gearhead and spindle).

Mechanical positioning accuracy
In this value, following factors are taken in consideration:
- backlash of the gearhead
- accuracy of the spindle
- axial play of the nut

Maximum stroke
The maximum possible stroke depends on the length of the spindle L [mm]. The length of the nut L_N [mm] and the thickness of its mounting plate X [mm] must be taken in consideration.

$$Stroke = L - (L_N + X + stroke\ reserve)$$



Mounting and safety instructions
Using a ball screw with a flange nut, the mounting through a hole is only possible with the optional rectangular mounting flange.

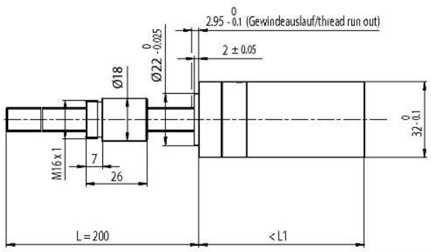
The ball screw nut may never be removed. As the balls are preloaded remounting would be impossible.

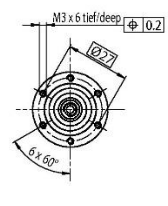
The spindle may never block during operation, as this could damage the spindle nut or gearhead.

APPENDIX A2 – THE SPINDLE DRIVE CATALOG
(GP 32 S ∅32 mm, METRIC SPINDLE)

Spindle Drive GP 32 S Ø32 mm, metric spindle

NEW





Options on page 8

Technical Data	
Spindle	M10 x 1, stainless steel
Option	M10 x 1, ceramic
Standard length	200 mm
Special length (5mm steps)	max. 600 mm
Max. efficiency	27.0%
Mass inertia	9 gcm^2
Nut (standard)	thread nut
Material	CuSn12
Axial play	< 0.008 mm
Planetary gearhead	straight teeth
Bearing	ball bearing / thrust roller bearing
Radial play, 5 mm from flange	< 0.05 mm
Axial play	preloaded
Recommended input speed [4]	< 8000 rpm
Recommended temperature range	-15 ... +80°C
Max. radial load, 15 mm from flange	200 N
Max. axial load (static) [1]	2700N

Stock program
Standard program
Special program (on request)

Order Number									
		363900	363904	363909	363910	363915	363920	363925	363930
Spindle Drive Data (provisional)									
1 Reduction		1 : 1	14 : 1	33 : 1	51 : 1	111 : 1	246 : 1	492 : 1	762 : 1
2 Reduction absolute		1/1	676/49	529/16	17576/343	13824/125	421824/1715	86112/175	19044/25
3 Max. feed velocity [1]	mm/s	100	9.5	4.0	2.6	1.2	0.5	0.3	0.2
4 Max. feed force (continuous) [1]	N	183	400	533	616	798	1040	1311	1350
5 Max. feed force (intermittent) [1]	N	455	995	1324	1350	1350	1350	1350	1350
Order Number		363901	363905		363911	363916	363921	363926	363931
1 Reduction		3.7 : 1	18 : 1		66 : 1	123 : 1	295 : 1	531 : 1	913: 1
2 Reduction absolute		26/7	624/35		16224/245	6877/56	101062/343	331776/625	36501/40
3 Max. feed velocity [1]	mm/s	36	7.4		2.0	1.1	0.5	0.3	0.1
4 Max. feed force (continuous) [1]	N	257	435		671	826	1105	1345	1350
5 Max. feed force (intermittent) [1]	N	638	1082		1350	1350	1350	1350	1350
Order Number		363902	363906		363912	363917	363922	363927	363932
1 Reduction		4.8 : 1	21 : 1		79 : 1	132 : 1	318 : 1	589 : 1	1093 : 1
2 Reduction absolute		24/5	299/14		3887/49	3312/25	389376/1225	20631/35	279841/256
3 Max. feed velocity [1]	mm/s	28	6.3		1.7	1.0	0.4	0.2	0.1
4 Max. feed force (continuous) [1]	N	280	458		712	845	1133	1350	1350
5 Max. feed force (intermittent) [1]	N	696	1139		1350	1350	1350	1350	1350
Order Number		363903	363907		363913	363918	363923	363928	
1 Reduction		5.8 : 1	23 : 1		86 : 1	159 : 1	411 : 1	636 : 1	
2 Reduction absolute		23/4	576/25		14976/175	1587/10	359424/875	79488/125	
3 Max. feed velocity [1]	mm/s	23	5.8		1.6	0.8	0.3	0.2	
4 Max. feed force (continuous) [1]	N	298	472		733	899	1234	1350	
5 Max. feed force (intermittent) [1]	N	742	1174		1350	1350	1350	1350	
Order Number			363908		363914	363919	363924	363929	
1 Reduction			28 : 1		103 : 1	190 : 1	456 : 1	706 : 1	
2 Reduction absolute			139/5		3588/35	12167/64	89401/196	15817/224	
3 Max. feed velocity [1]	mm/s		4.8		1.3	0.7	0.3	0.2	
4 Max. feed force (continuous) [1]	N		504		778	955	1278	1350	
5 Max. feed force (intermittent) [1]	N		1253		1350	1350	1350	1350	
6 Number of stages		1	2	2	3	3	4	4	4
7 Max. efficiency gearhead incl. spindle	%	22[2]	20	20	19	19	16	16	16
8 Weight [1]	g	304	331	331	359	359	387	387	387
9 Average backlash no load	°	0.7	0.8	0.8	1.0	1.0	1.0	1.0	1.0
10 Mechanical positioning accuracy [1]	mm	0.033	0.033	0.033	0.034	0.034	0.034	0.034	0.034
11 Mass inertia gearhead incl. spindle [1]	gcm^2	3.0	0.9	0.9	0.7	0.7	0.7	0.7	0.7
12 Gearhead length L1	mm	50.2	56.9	56.9	63.6	63.6	70.3	70.3	70.3

[1] based on Spindle length 200 mm (standard length) [2] for reduction 1:1 = 27% [3] for reduction 1:1 = 43.3 gcm^2 [4] for reduction 1:1 = 4000 rpm

Combination											
+ Motor	Page	+ Tacho / Brake	Page	Overall length [mm] = Motor length + gearhead length + (tacho / brake)							
RE 35, 90 W	82	DCT22	271	139.3	146.0	146.0	152.7	152.7	159.4	159.4	159.4
RE 35, 90 W	82	AB 28	308	157.3	164.0	164.0	170.7	170.7	177.4	177.4	177.4
RE 35, 90 W	82	HEDS 5540' /	262/308	174.4	181.1	181.1	187.8	187.8	194.5	194.5	194.5
RE 36, 70 W	83			121.5	128.2	128.2	134.9	134.9	141.6	141.6	141.6
RE 36, 70 W	83	MR	259	132.9	139.6	139.6	146.3	146.3	153.0	153.0	153.0
RE 36, 70 W	83	HED_ 5540	262/264	142.5	149.2	149.2	155.9	155.9	162.6	162.6	162.6
RE 36, 70 W	83	DCT22	271	139.6	146.3	146.3	153.0	153.0	159.7	159.7	159.7
A-max 26	115-122			95.0	101.7	101.7	108.4	108.4	115.1	115.1	115.1
A-max 26	115-121		270	102.1	108.8	108.8	115.5	115.5	122.2	122.2	122.2
A-max 26	116-122		254	103.8	110.5	110.5	117.2	117.2	123.9	123.9	123.9
A-max 26	116-122		260	109.4	116.1	116.1	122.8	122.8	129.5	129.5	129.5
A-max 26	116-122		262/264	113.8	120.5	120.5	127.2	127.2	133.9	133.9	133.9
A-max 32	123/125			113.2	119.9	119.9	126.6	126.6	133.3	133.3	133.3
A-max 32	124/126			111.8	118.5	118.5	125.2	125.2	131.9	131.9	131.9
A-max 32	124/126		255	123.0	129.7	129.7	136.4	136.4	143.1	143.1	143.1
A-max 32	124/126		262/264	132.6	139.3	139.3	146.0	146.0	152.7	152.7	152.7
RE-max 29	145-148			95.0	101.7	101.7	108.4	108.4	115.1	115.1	115.1
RE-max 29	146/148		254	103.8	110.5	110.5	117.2	117.2	123.9	123.9	123.9
EC 32, 80 W	164			110.3	117.0	117.0	123.7	123.7	130.4	130.4	130.4
EC 32, 80 W	164	HED_ 5540	262/264	128.7	135.4	135.4	142.1	142.1	148.8	148.8	148.8
EC 32, 80 W	164	Res 26	272	130.4	137.1	137.1	143.8	143.8	150.5	150.5	150.5

6 maxon Spindle Drive

More combinations on page 5 and 7.