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RUHR-UNIVERSITÄT BOCHUM



Vietnamese-German University

MechEng
Mechanical Engineering
INTEGRATION

DESIGN, FABRICATION AND TESTING OF PROSTHETIC LEG

BACHELOR THESIS

HO CHI MINH 2023

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DESIGN, FABRICATION AND TESTING OF PROSTHETIC LEG

A Thesis presented

By

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Vietnamese-German University

Submitted to the department of Mechanical Engineering of the

RUHR-UNIVERSITÄT BOCHUM and VIETNAMESE-GERMAN UNIVERSITY

in partial fulfillment

Of the requirement for the degree of

BACHELOR IN MECHANICAL ENGINEERING

February 2023

Major: Mechanical Engineering

VGU
Vietnamese-German University
NGUYEN THAI BINH

MEN 2017

DESIGN, FABRICATION AND TESTING OF PROSTHETIC LEG



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Abstract

This thesis produces to fulfill the requirements for the Mechanical Engineering Bachelor's Degree at the Vietnamese-German University.

The goal of this project is to design, fabricate, and testing of a prosthetic leg to create a prosthetic. The Leg design was initially built as an initial model, followed by optimization in Topology Study, and then rebuilt as a validation model using the graphical reference provided by the Topology Study's outcome. In Solidworks, CAD modeling is created. Based on a structural statistics study of the design carried out by utilizing Finite Element Analysis (FEA) to comprehend the stress and displacement distribution throughout the designs. A prosthetic Leg is usually costly and uses expensive material. This project will emphasize this problem of the prosthetic legs by focusing on a design that uses simple engineering mechanics and alternative materials for knee prosthetics.

The results of this research are also the comparability of the characteristics of the present and proposed materials, which help to provide substitute materials that are more economical and environmentally friendly while maintaining the features needed for prosthetic limbs. The research is intended to improve patient access to more economical products and advance prosthetic leg design.

AUTHORSHIP STATEMENT

This project was accomplished by two MEN2017 students of the Vietnamese-German University.

Nguyen Thai Binh and Nguyen Anh Vu are combining their work as a thesis project for the degree of Bachelor in Mechanical Engineering program.

Each team member contributes to various parts of all components of the project.

I hereby certify my authorship of the bachelor thesis submitted today entitled: " DESIGN, FABRICATION, AND TESTING OF THE PROSTHETIC LEG ".

The result of my own research, except where otherwise acknowledged, and that this thesis has not been submitted for a higher degree at any other institution. To the best of my comprehension. The thesis contains no information previously published by other individuals except when the reference is given in the thesis itself.

Author's signature

Chapter 1: Introduction

Research Background and motivation

The development of science and technology in the fields of human health

We have entered the 21st century, known as the century of rapid science and technology advancements. In this age, the achievements of science and technology are applied to all aspects of life, not limited to the industrial and manufacturing sectors/environment. In daily life, we can see the presence of machines everywhere. In transportation, many have been built to meet the demand for global freight (tracking and monitoring the location of containers)

In household applications, many advanced security systems and automation systems (lighting, monitoring, health tracking, locking doors) can be controlled using a phone, all within our grasp. However, the need of man is infinite, and it arises under certain circumstances. Hence conventional technology can only meet users' needs for a period of time before being further developed. Thanks to the continuous development of technology. Especially in the health sector, countless types of research have been done all over the globe to increase life expectancy and living conditions for patients of all departments in a hospital. As a matter of fact, machines and technologies are developed by humans to serve the needs of our life's problems. In that direction, the development of science and technology in human health unstop growing. With today's development rate, the public's health has improved significantly compared to the 20th century. Life expectancy has increased, mortality rate caused by the disease has decreased. Incurable diseases in the past can now be solved through the development of breakthroughs in vaccines, antibiotics, etc. Thanks to the development of Modern technology and equipment, the treatment of patients has greatly improved: High-precision robotic arms have gradually replaced/are slowly taking the places of human hands in surgery, and they can perform complex surgical operations with broader access to internal organs of the body with only a small incision instead of having to cut a large part like when humans do => improve safety and post-surgery recovery. The appearance of CT machines, PET/CT machines, EEG machines, MRIs, and ultrasounds help doctors diagnose diseases more precisely than ever and shorten diagnosis time.

With the development of technology, even today, we can replace lost body parts with all types of prosthetics that integrate with MR damper, creating suspension to simulate real human, polycentric using mechanism system simulate human movement. Also, the development of 3D printers makes more people get more suitable Prosthetics, for example, flexible arm and leg parts. The increase of available material due and human biological studies.

Due to the characteristics of the field of study as well as the requirements of society, this led to one question, why do not use this based available development to make a design for prosthetics using this technology?

Unsurprisingly, this has, in fact, already been done, and not even limited to the mechanic field but it also involves computerized with complex designs but it also that most medical prostheses for amputees on the market are quite limited in terms of design and materials, as well as relatively high prices, and are less suitable for developing countries with the majority of the population being working people. The inability to pay for a Prosthetic leg because of the expenses leads to amputees in these countries being unable to travel, creating a burden on themselves, their families, and society.

This thesis provides more data and analysis for the integration and development of future prosthetics, which can be beneficial for public health development.

Motivation

Today, the primary reason for amputations is dysvascularity, a condition that frequently results from or connects to diabetes [1]. While rising in demand, the supply of prosthetic devices is still limited. Engineers are needed to design better limb replacements. The purpose of us as an engineer is to create products that perform better than the products with the same price on the market. The aim of this study is to find the right design of a prosthetic leg for the developing country. Creating a design focus on mechanisms and using alternative materials that are more cost-efficient and eco-friendly, easy to maintain for developing countries but still keep the attributes necessary for prosthetic limbs. The findings assist patients with a cheaper product and approach the development of the advanced prosthetic limb. With The topic "Design fabricates and testing of an affordable prosthetic leg". I hope to contribute to the advancements of the medical industry as well as human health in the future.

Chapter 2: Literature review

Introduction

This chapter provides an overview of literature expressing the fundamentals and state of research of prosthetics as well as leg anatomy functional to find the basic requirement of our design before we go deeper into each part. Analyzed data in this review are four primary areas: Prosthetic History, Leg Functional Anatomy, Human Movement, and Knee Prosthetic.

Background study/fundamental and state of research

Prosthetic History

As a result of World War II, the development of prosthetic devices surged considerably as a result of the enormous amount of amputees after the war. The majority of those cases were young people who needed the most cutting-edge prosthetic technology to regain their function. Engineers had a significant and enduring impact on the area at the time, and they are still actively involved in prosthetic design today to transform the industry. Probably the most important development was the consequence of procedure design cooperation between engineers and doctors in the surgery field for the prosthesis's best alignment and fitting, and optimal restoration of ambulation [1].

Leg Functional Anatomy

So how can we, engineer can understand the prosthesis field, first we need to know the principles and focus on the parts we need to restore in this prosthetic leg design.

Let us see how the lower leg movements are show in **Figure 1**. The quadriceps group include the hips, knee, and ankle, the three main joints of the human leg and the movement of them.

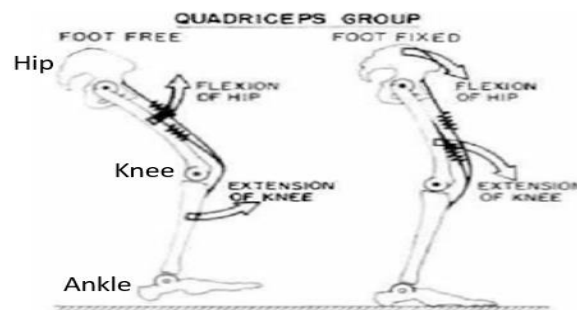


Figure 1 Quadriceps group [1]

Flexion-extension of the hip and the knee and horizontal rotation are the three axes in which the hip can spin. By combining these rotations, the hip can also be circumduction. Fundamentally, the hip is a ball and socket joint. The knee slips while rotating along the condyle, bending and extending like a hinge in one plane (sagittal), it operates as a "four-bar" mechanism [1].

After knowing the movement of the leg the next thing is find out how the leg work **Figure 2.a** represent the main muscle of human leg Femoris and Vastus, gastrocnemius and soleus.

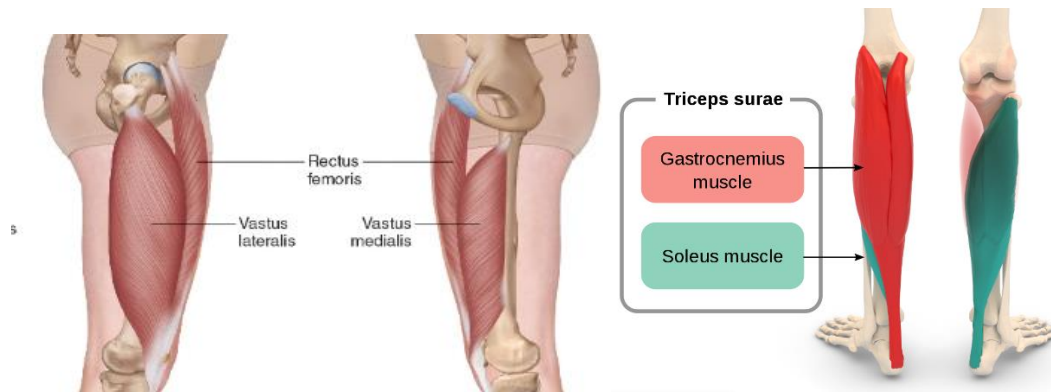


Figure 2.a: The muscles of the lower leg (source internet)



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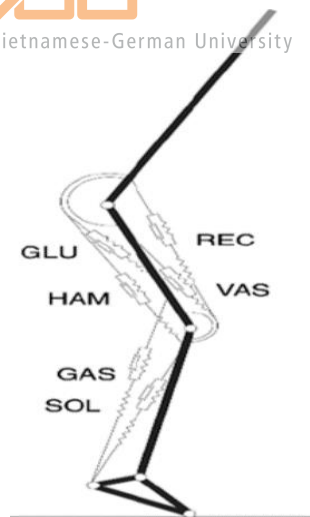


Figure 2.b: The muscles of the lower leg [1]

For easy for design the lower leg's muscles are represent as a spring and dashpot and shown in **Figure 2.b**. The Rectus Femoris and Vastus muscles, denoted by the letters REC and VAS, are the two main knee flexors. These muscles make up the quadriceps, which attach to the tibia and fibula at sites immediately behind the knee joint and extend downward from the

hip. The hamstrings extend the knee, the gastrocnemius and soleus flex the ankle, and the gluteus muscles flex the hip. Each muscle is modeled as an active contractile element connected to a spring or dashpot in series. The action occurs simultaneously to make a leg move. So to recreate the leg movement we have to strictly stick on this study and transform this into the type of suitable mechanic to create the same function.

Human movement

Gait Cycle

Next, in order to understand leg trajectory, analyze how human move their legs are needed. The gait cycle is taken into consideration to fully understand how the human body controls the lower body to advance while maintaining balance.

As shown in **Figure 3** below, the movement of our leg consists of many continuous phases with each phase is a combination of flexion and extension of different muscles within the leg.

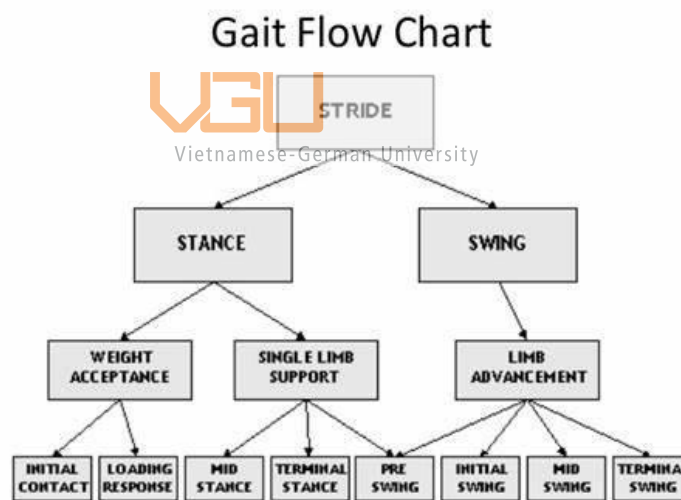


Figure 3 Human gait Flow chart [2]

Gait cycle can be categorized into 2 predominant phase: Stance phase and Swing phase. However, if we take in detail, total of 6 micro phases is determined in the gait cycle: heel contact (HC), foot flat (FF), mid-stance (MST), heel-off (HO), toe-off (TO) and mid-swing (MSW) [2] as shown in **Figure 4**

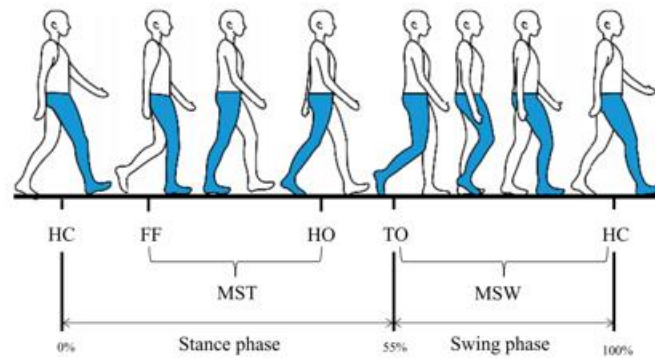


Figure 4 Classification of the human gait cycle [3]

The gait cycle of one leg begins at the heel contact, the leg starts reaching out to initiate the new step, with the heel making contact to the ground, next, the foot is positioned flat to the ground surface, at this position the leg acts as a force absorber and balancer for the leg, notice at this position the leg can withstand maximum weight. Mid stance is the transition phase where from flat foot to initiate heel off and toe off leading to the swing phase of the foot (mid-swing) [3]

The stance phase takes up about 60% of the gait cycle, and the swing phase takes up around 40% [3]. One factor to be considered when mentioning the safety of the human leg when walking is Toe clearance [3]. **Figure 5** illustrates the heel and toe clearance pattern during gait. Toe clearance is chosen because it is used to measure the vertical distance from the ground to the lowest point of the toe during the walking cycle. If that obstacle is higher than the toe clearance, the prosthetic user cannot walk past that and may cause falling. Toe clearance, the primary performance factor that we need to consider and improve, happens with the main major index in the swing phase. During the swing phase, we defined the toe clearance by the minimum of the distance vertical of the lowest point of the leg while swinging with the ground or floor surface. [4]

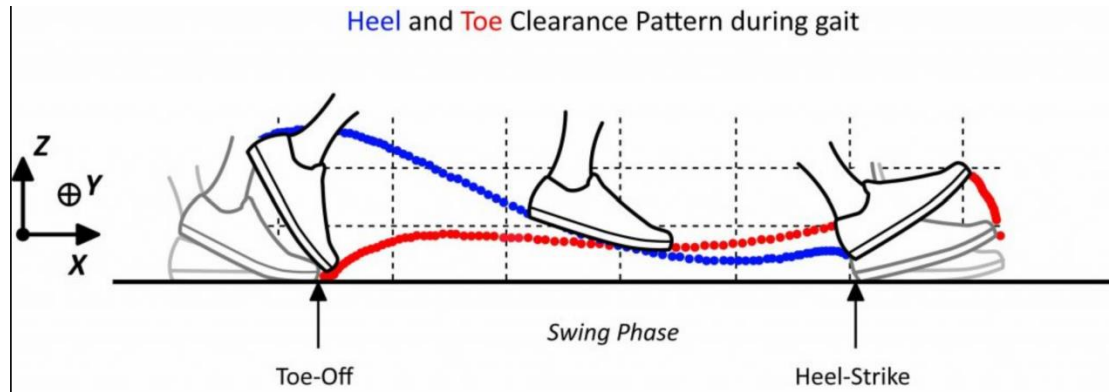


Figure 5 Heel and Toe clearance during gait [5]

When met with an obstacle, the patient can raise the leg simply by lifting their remaining thigh. However, performing many times can lead to accumulated stress on the hip, and make it difficult to work for a longer period of time. A healthy toe clearance of an able-bodied person is ranged from 10 to 20 mm from the surface [3]. To test the toe clearance tester using a visual test where LED lights are attached to our prosthetic leg at the 3 main joints: hip, knee angle, and at the lower point of our toe to estimate the minimum toe clearance [5].

Types of Prosthetic Knees joints

In this Part study, we look at the type of knee joint available in the market, and its advantage and disadvantage, between mechanical or computerized prosthetic knees. Mechanical Knees are still the main focus since this project aims to create a prosthetic knee suitable for developing countries where most of the population is working class. So the price point must be considered.

Every prosthetic knee needs a stability mechanism, whether a manual locking system or one actuated by weight. They also require a method for directing the movements of flexion and extension. This may be done via friction or hydraulic/pneumatic control. [6].

Above knee prosthetic leg (AK type)

Here prosthetic chosen to study is the AK (Above knee) since most of the movement function is lost, which means that the amputated position is located above the knee, and a part of the leg is just retained from the hip. Our leg design process will mainly focus on hip control, the structure of the knee as well as the support structure of the lower leg. [1]

Single-Axis Knees

Mechanism: a simple hinge knee shows in **Figure 6**.

These structures carry out a straightforward rotation around the knee axis during flexion and extension. Single-Axis knees have a basic design, and easy alignment complies with mechanical rules. Single-Axis knees are available with manual or automated blocking of flexion for use by people with weak muscles. [6]



Figure 6 Single-Axis Knees [6]

Advantages: straightforward, robust, light, and affordable.

Disadvantages: Due to the simplicity, the person must utilize their own muscular force in the limb to maintain the knee's stability when standing with the heel in touch. Because there is minimal to no hip support, walking for a long period of time can lead to muscle fatigue.

Weight-Activated Stance Control also called "safety knees" (Kind of Single-Axis Knees)

Mechanism: The knee has a constant friction mechanism, which means that while the patient puts weight on the prosthesis, it will apply a braking force to keep the knee from buckling. The knee will swing freely all of the time until weight is placed on it [6] show in **Figure 7**.

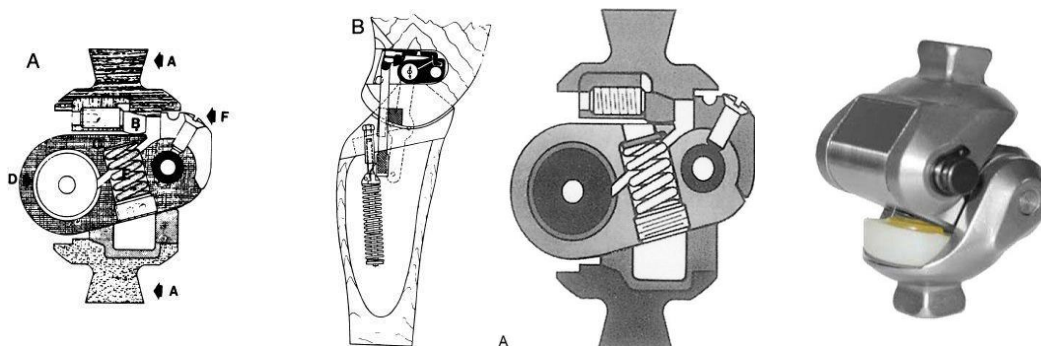


Figure 7 Single-Axis Knees [6]

Advantages: Stable knee. Primarily prescribed for first-time prosthetic users who require stability, particularly in the elderly or less active population, but who can still exercise some control over the knee. Or someone who becomes exhausted after just a few steps. This is especially helpful for patients who accidentally put weight on a knee that is only half bent since the friction in the knee will brake and stop the knee from falling into flexion.

Disadvantages: The patient will be unable to utilize their prosthetic side when sitting down since they must remove their weight from the leg in order for it to flex the patient will walk more slowly and take smaller steps as a result of the friction in their knee. [6]

Manual Locking Knee

Mechanism: The most stable knee utilized in prosthetics is the manual locking knee. During walking, the knee is locked, but the lock is released when it's time to sit down. Patients who are unable to manipulate their knees due to having very short residual limbs or weak hips typically employ manual locking knees [6] show in **Figure 8**

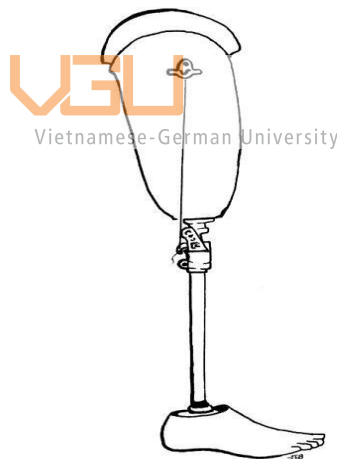


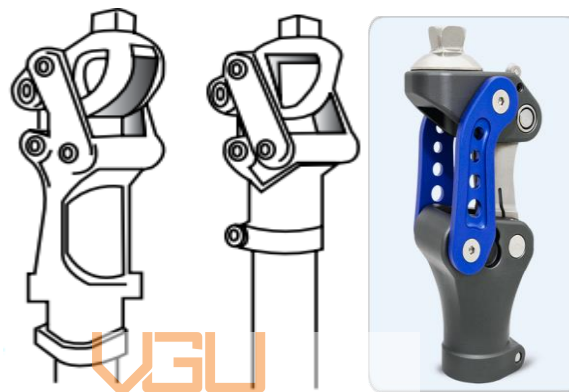
Figure 8 Manual Locking Knee [6]

Advantages: This will enable for locking the knee with weight bearing the patient could decide to manually locking their knee. This is excellent for persons who require more protection to prevent the knee from mobile or disabled individuals who cannot freely control their own prosthetic knee. Extremely straightforward, robust, light, and affordable.

Disadvantages: the locked knee requires excessive energy to use and produces a stiff, awkward gait [6]

Polycentric Knees

Mechanism: Polycentric Knees This knee has several axes of rotation. Polycentric knees may be four-bar knees (4 axes of rotation) or seven-bar (seven axis of rotation). The most common usage is of 4 axes (or 4 bars), Their flexibility is the main factor behind their setup, Polycentric Knees extremely stable in the early stance phase, simple to bend to begin the swings phase or sitting, another popular attribute of the knee's design is that the total length of the leg shortens as soon as a step is taken, lowering the possibility of stumbling. Polycentric knees can be used for a diverse group of amputees. Various models are perfect for amputees who cannot secure the other knees when walking, have bilateral or knee disarticulation amputations of the legs, or have lengthy residual limbs. [6] show in **Figure 9**



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Figure 9 Polycentric Knees [6]

Advantages: It is incredibly adaptable in terms of stability and can be set to be very stable when the patient enters the stance phase, but at the same time it also permits an effortless swing and allows sitting down with a bent knee. The various axes and instant Centre of Rotation (ICR) show in **Figure 10** cause the prosthesis length to "shorten" at the beginning of the toe-off, allowing for foot clearance.

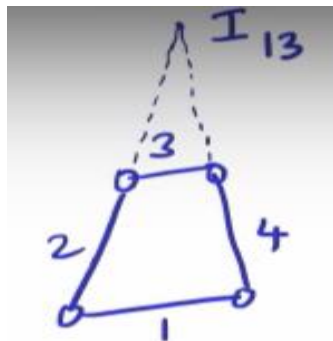


Figure 10 Instant Centre of Rotation of link 3 with respect to link 1

Cons: It is only for a single walking speed, and weighs more than a knee with one axis. more components that require maintenance and replacement [6].

Pneumatic or Hydraulic Knees

Mechanism: When a single or poly-axial knee is supplemented with a pneumatic or hydraulic component show in **Figure 11**.

These parts enable the user to change speeds while controlling the prosthetic knee's swinging motion. When the speed is raised, the cylinder's valve shuts and progressively reduces the airflow or fluid pressure to restrict knee flexion. When knee flexion is limited, the person can walk more quickly.



Figure 11: Pneumatic or Hydraulic Knees [6]

Advantages: A more natural posture will enable the person to walk more easily at various walking rates. When sitting down, weight can be kept on the prosthetic leg, and the knee will help the person get comfortable. the resistance in the knee will allow the individual to climb down step over step when walking downstairs.

Disadvantages: Compared to pneumatic knees, hydraulic knees are more expensive initially, heavier, and require more care. The whole price is more than the knees discussed before [6].

Computerized Knees

Mechanism: The sensors inside the knee joint and/or the foot provide feedback to a microprocessor in these knees. The sensors' data is utilized to modify the pace and range of knee flexion and extension to suit the individual's needs at the time [6] show in **Figure 12**.



Figure 12 Computerized Knees [6]

Advantages: requires less energy and effort when walking thanks to the microprocessor knee, resulting in a more natural gait. the knee can swiftly adjust to varied walking speeds, a changing environment, or certain circumstances. Some knees include a stumble recovery feature as well, which helps keep people from falling and can be modified using a PC or a mobile device.

Disadvantages: Extremely high cost. The battery requires recharging. More weight than other knees. Designers could be challenging specific choice of the foot, environmental factors (water, heat, cold, etc.), kneeling, and other factors can harm it [6].

Chapter summary

By this chapter, we have obtained knowledge of the movement of the human leg, its function, gait cycle, toe clearance, and how it affects walking. We also study the prosthetic leg available on the market and the advantage and disadvantages of each. The form that we can conclude in the basic requirement of our design, an appropriate way to build a polycentric knee that satisfies the demands of physical movements and re-create the gait cycle of the leg for users in developing nations, where stability and toe clearance is crucial because of the rough terrain faced.

Optimization consideration: hip flexion and extension moments, alternative material and price point.

This can be sum up into 3 major aspects for the requirement for our design:

- The design material and price point (Price point)
- Ability to provide sustainability in standing and support when walking (Natural posture, Swing support)
- Ability to Increase the ability user use for walking on uneven terrain (Toe clearance).

Chapter 3: Conceptual design

Conceptual design

The mapping draft in **Figure 13** below show the conceptual design of the polycentric knees from how to select the parametric to the actual product finished

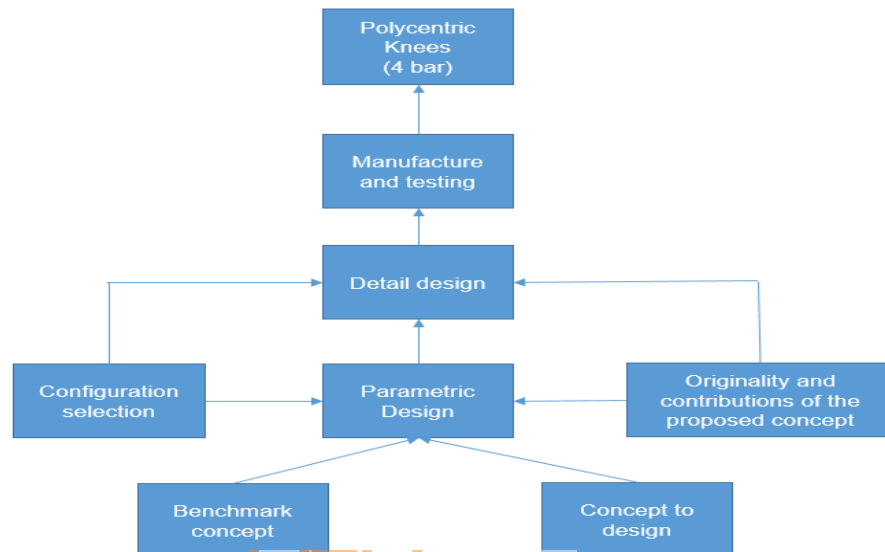


Figure 13 The general methodology applied

Configuration selection

In this part, in order to choose the design that is reasonable with the requirements, we have used a score on a scale of 1 -5 for each category, from which to choose the most optimal design. Our goal is to make a knee that not only can achieve the full function of a lower Limb Prosthetic but is also easy to use at a price that is suitable for developing countries.


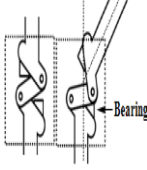

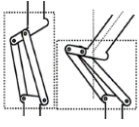

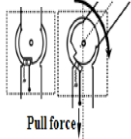

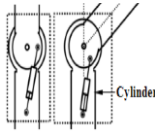

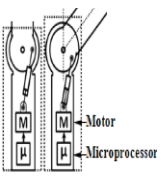
From different knee types, we find out the most suitable knee requirements for an AK. Prosthetic Knee:

- Provide stainability in standing and walking
- Shock absorption
- Enable to forward motion of the leg for the next step
- Control pendular swing
- Reduce effort to walk
- Restore independent

The concept scoring must have adapted both role of the knee prosthetic and role of requirement of our design, so the scoring will have a category that we focus on in blue words. The list of scoring category that based on the role of requirement that make our design difference:

- The design material and price point (Price point)
- Ability to provide stainability in standing and support when walking (Natural posture, Swing support)
- Ability to Increase the ability user use for walking on uneven terrain (Toe clearance)

The **table 1** below will provides the concept scoring and what main category of the design

Knee design	Schematic	Range (degree)	Weight	Price point	Toe clearance	Maintenance and Production	Natural posture	Swing support	Siting comfort	Total
		0-150	5	5	1	5	1	1	1	19
Polycentric Knees (4 bar) 		0-160	4	4	3	5	3	3	3	24
Manual Locking Knee 		0-140	4	5	1	5	1	1	2	19
Pneumatic or Hydraulic Knees 		0-160	2	1	4	2	4	4	5	23
Computerized Knees 		0-160	2	1	5	1	5	5	5	24

Range point 1-5 (Bad-Good)

Table 1: concept scoring table

As compared to the **table 1**, we can easily see that the knee with a 4-bar structure and the Computerized Knees have the highest concept scoring. In the table of requirement points (blue words), the 4 bar and the Computerized Knees also have the same highest point, so the price point has to take into account to compare and choose one suitable design only.

Due to the requirement of price point, the Computerized knee is neglected. ⇒Therefore, the Polycentric 4-bar mechanism was selected.

Summary

After this part following the scoring table we have cleared one thing, the main focus of design and design categories from now will be the polycentric 4-bar mechanism prosthetic knee.

Concept to design

After choosing the suitable design, we need to study the 4-bar mechanism. To begin, we focus on analyzing the mechanism of the 4-bar and compare it with the biomechanics of humans with the purpose to re-create the same biomechanics. We also apply a heuristic search method to develop a suitable 4-bar design with optimal values for the given parameters that largely concentrate on the required design.

- The design material and price point (Price point)
- Ability to provide sustainability in standing and support when walking (Natural posture, Swing support)
- Ability to Increase the ability user use for walking on uneven terrain (Toe clearance)

This design will also help us to reduce the price due to the simplicity and ease of Maintenance.

Biomechanics of human knee represent in 4 bar mechanism

First, we need to understand the structure of the 4-bars and its components and dynamic structure of the 4 bars as well as the symbols used.

The 4 bar is computed by link and joint:

- Links: rigid member having nodes
- Links: a rigid member of any shape having a number of attachment call nodes, linkage are connected specifically to have control output

Binary link: a link that has 2 nodes.

Ternary link: a link that has 3 nodes.

Quaternary link: a link that has 4 nodes.

- Node: attachment points
- Joint: connection between two or more links (at their nodes) which allows motion

In the design of the 4 bar prosthetic for the optimize of material the Binary link is use

Link classification:

- Ground: fixed w.r.t. reference frame
- Crank: pivoted to ground, makes complete revolutions
- Rocker: pivoted to ground, has oscillatory motion
- Coupler: link has complex motion, not attached to ground

The 4 bar prophetic knee is denoted as RRRR, constructed from four links connected by four revolute joints. The planar quadrilateral linkage refers to all arrangements in this type.

Type of Linkage:

4R linkages include:

- Double-crank linkage
- Crank-rocker linkage
- Double-rocker linkage
- Parallelogram (Parallel Motion)

With actual knee movement from Biomechanics of human knee represent in 4 bar mechanism show **in figure 14**

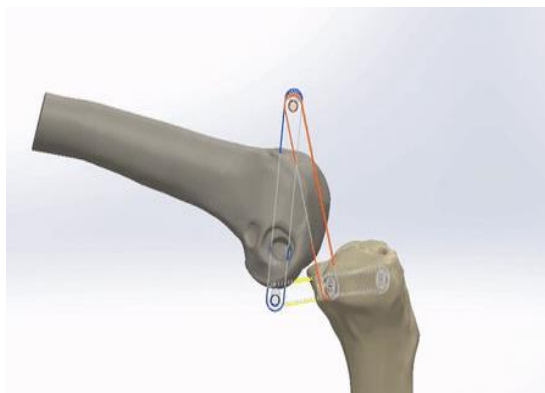


Figure 14 Biomechanics of human knee represent in 4 bar mechanism

Compared to the Biomechanics of the actual knee in **figure 14**, it can be seen as one link controlling another movement with a rotation of about 120 degrees. It can be understood that the Crank Rocker system is chosen.

Degree of freedom: Number of independent parameters that define its configuration. It is the number of parameters that defines the state of a physical system. To know the knee work that is similar to the human knee, first, we need to determine its DOF. The Prosthetic knee has to be able to do two main phases of the normal human knee *the swing phase and the stance phase*. The 4-bar knee is capable of flexion and extension; this motion of the knee requires 1 DOF that allows rotation in an x-y coordinate.

In this part, to find out the DOF we use Gruebler's equation for planar mechanisms

$$M=3L-2J-3G \Rightarrow M= 3(L-1) - 2J_1 - J_2 \quad (3.1)$$

Where

M = degree of freedom or mobility

L = number of links

J = number of full joints (half joints count as 0.5)

G = number of grounded links =1

Here with the 4 bars mechanism we calculate the moment of the Knee

Cause the knee have to produce the flexion and extension of the knee, the knee requires 1 DOF that one movement. **Figure 15** show the 4-bar prosthetic knee represent in term of links and joints.

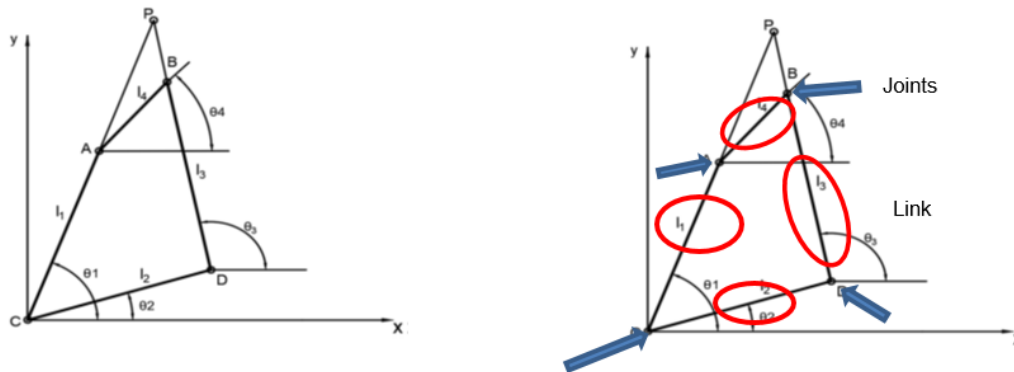


Figure 15 Human knee represent in 4 bar mechanism (source internet)

Number of links = 4

Number of 1 DOF joint = 4 (4 Revolute)

$$M= 3(L-1) - 2J_1 - J_2$$

$$M= 3(4-1) - 2(4) - 0$$

$$M = 9 - 8 - 0$$

$$M = 1$$

With $M=1$ the 4-bar Polycentric has 1 type of motion that is flexion and extension in swing phase of the leg.

Applying the Grashof Rule, we calculate the rotation limit to design the linkage, Due to the similarity of the human knee. The motion of the knee is commonly modeled as a planar four-bar mechanism **figure 16** represent Human knee represent in 4 bar mechanism

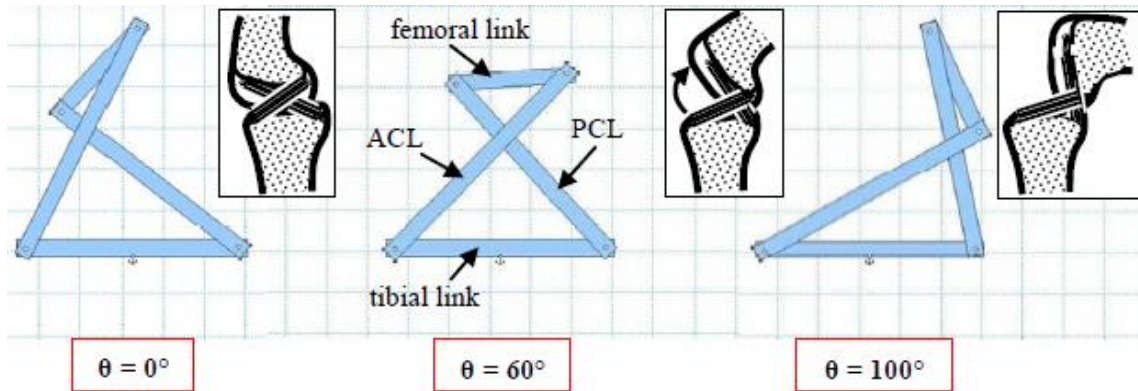


Figure 16 Human knee represent in 4 bar mechanism [7]

As we can see from **figure 16** Human knee joints locks in an upright position. understand that 1 link must be crank

Grashof condition predicts behavior of linkage based only on length of links

$$S + L > P + Q \quad (3.2)$$

S=length of shortest link

L=length of longest link

P,Q=length of two remaining links

The Grashof's Rule stated that :

- If $S + L \leq P + Q$ the linkage is Grashof :at least one link is capable of making a complete revolution
- Otherwise the linkage is non-Grashof : no link is capable of making a complete revolution

To make the knee have same function of the human knee the knee must be Non-Grashof (**1 link must be crank**)

So the design knee must be Non-Grashof have condition $S+L>P+Q$.

From the study of 4 bar and human knee mechanic of design must have the following condition to be fully function as a knee replacement prosthetic:

- Design use Binary link
- Design is a Crank Rocker
- Design must have 1 degree of freedom
- Design knee must be Non-Grashof ($S+L>P+Q$)

Benchmark concept

After choosing the suitable design condition, we focus on analyzing how the previous polycentric 4-bar was designed and what category needs to consider when designing a 4-bar prosthetic knee. In **figure 17** below, we take a look at past, and present work, the mechanism of prosthetic knee 3R20 made from Stainless Steel was examined and studied. Looking at this past work, we find what category needs to consider when designing a 4-bar prosthetic knee. This is a 4-bar type of knee that has the aesthetic knee center positioned at a height of 500mm. The hip is placed 450mm above the knee centroid, straight above the ankle. The heel is 50mm posterior to the alignment reference line (ARL: the line uniting the hip and ankle), while the toe lies 200mm in the forefront of the ARL. With corresponding edge parameters.

[7]

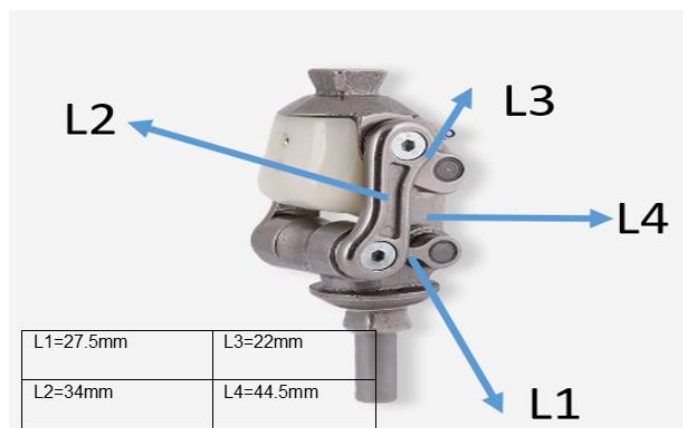


Figure 17 Prosthetic knee 3R20 made from Stainless Steel (Ottobock)

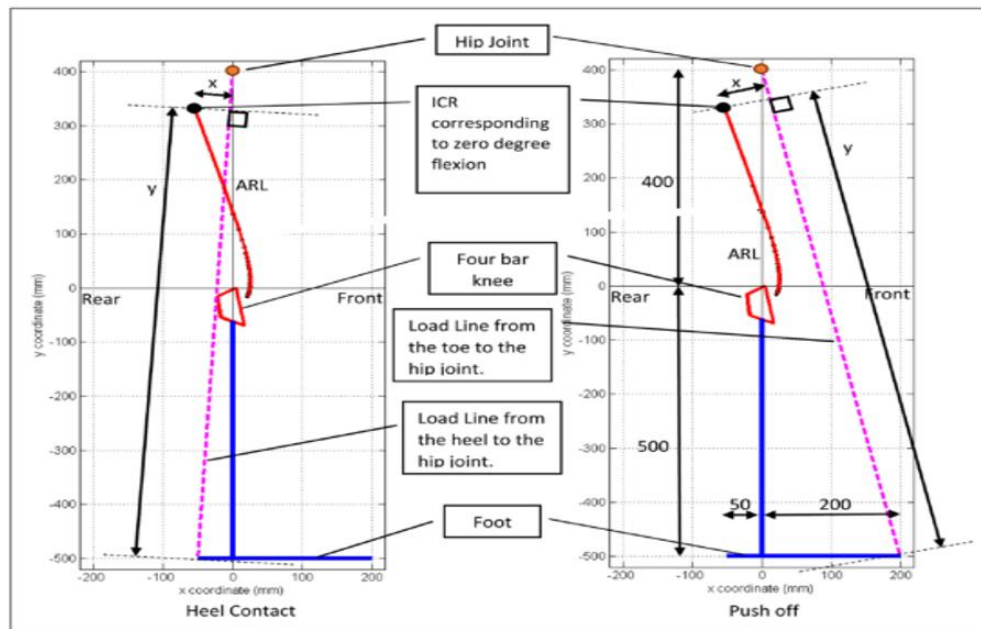


Figure 1. Calculation of x/y ratios for a typical four-bar knee.

Figure 18 component of four-bar knee [7]

Following **figure 18** and the research paper of prosthetic knee 3R20 the design of four-bar prosthetic knees can condensation into major focus points:

- The localization of the instant center (ICR) is defined by the junction of the front and back links elongating. This ICR is created to obtain a suitable polycentric mechanism that the amputee can fully control and estimate the knee angle of the prosthetic that can perform up to design.
- Load line (The reaction force exerted between the foot and the floor acts along a line) is the line linking the hip joint with the center of pressure on the foot. This line is the line of the body's "center of gravity", which the net reaction force exerted between the foot and the floor activates. Load line is one of the main focuses to be considered when designing because it directly affects the stability of movement.
- The x/y ratio in **Figure 18** is calculated by the position of the instant center of rotation with respect to the load line. The location front or behind the load line is determined via x , and the height of the knee center point above the floor measured along the load line is y the x/y ratio of a polycentric knee relies on the geometry of the four-bar knee and changes with knee flexion. This x/y ratio divides into a 2-part x/y ratio at heel contact and an x/y ratio at push-

off. The heel contact x/y ratio is important for stability at the stance phase, and the push-off x/y ratio is important for stability at the swing phase.

- Toe clearance: the displacement of the ground and the toe during the swing. This directly affects the ability to walk on uneven terrain and obstacle. So the design must be optimized for this index to have better stability on uneven terrain. [7]

Summary

This part presented the previous design of 4 bar knee and category need to consider when design a 4 bar prosthetic knee by this study the main focus when design a 4 bar is:

- The ICR trajectory
- the aesthetic knee center position based on Load line
- x/y ratio of stabilize
- The toe clearance
- And one more aspect that need to consider is material differential

Originality and contributions of the proposed concept

Design criterion and point of differential:

After studying the previous design and finding the condition to make the design fully functional, we now focus on the requirement of the design that makes our design more optimized compared to the previous design.

1. In order to differentiate our product from its predecessors we optimize the design our design target was a knee mechanism with the following features:
 - Ability to provide stability in standing and support when walking (Natural posture, Swing support)
 - Ability to Increase the ability user use for walking on uneven terrain (Toe clearance)
 - The design material and price point (Price point)
2. To make the design that satisfies the requirements the following point in design need to be done:
 - Ability to provide stability in standing and support when walking (Natural posture, Swing support) => Stable and effortless heel contact and load, which

the ICR reposition for stability, and a create a supported force for initial swing. (1st)

- Ability to Increase the ability user use for walking on uneven terrain (Toe clearance) => Maximized toe clearance during swing. (2nd)
- The design material and price point (Price point) => Optimize the material selection (3rd)

Point 1: Ability to provide stainability in standing and support when walking (Natural posture, Swing support)

Solution: Stable and effortless heel contact and load, which the ICR reposition for stability, and create a supported force for initial swing. (1st)

To solve the requirement, we have combined two structures commonly used in prosthetic legs, between the traditional 4 bars with the central cut of the single-axis prosthetic so that the natural support mechanism can be activated without the need for external support (spring or damper) thereby reducing the cost as well as the cost of manufacturing. To better understand the centering of a single-axis prosthetic leg we first learn the theory of load line and zone of stability.



Load line

Definitions for the lower-extremity prosthesis. The lower leg of the prosthesis includes 3 components: center of gravity, weight line, and load line. **Figure 19 illustrates the normal human load line.**

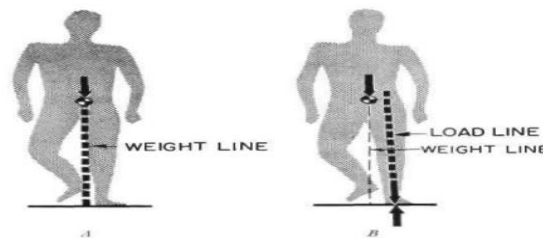


Figure 19 Human load line [7]

The body's "center of gravity" is the location where it is supposed that all of its weight is concentrated.

The effective bodyweight moves vertically downward along the "weight line" after passing through the center of gravity. The force exerted between the foot and the floor acts along a

line known as the "load line." Since this force has two effects, it is typically not perpendicular to the floor surface. As we can see, it does two things: it supports the body weight in a vertical direction and it generates the horizontal forces required to move the body forward and medially. The load line plays a crucial role in how a human walks. [8]

The principle of load line

Figure 20 represent Stage of movement when walking and he zone of voluntary stable of load line

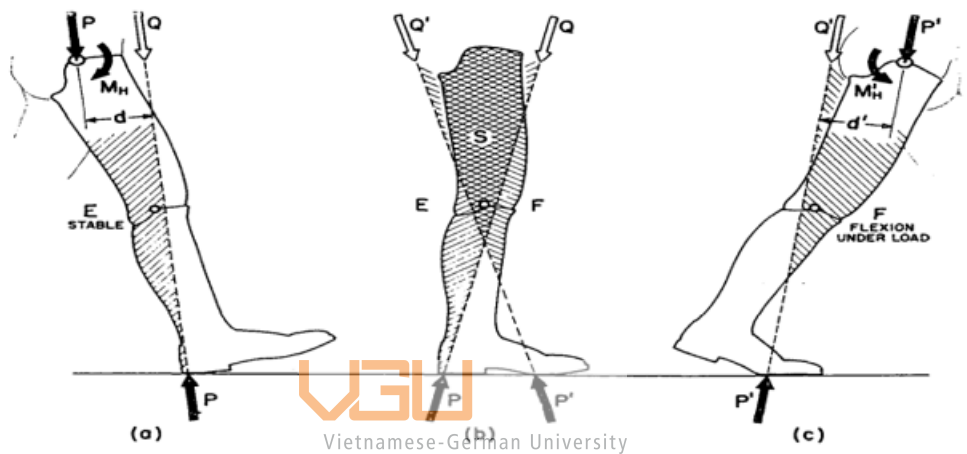


Figure 20 Stage of movement and he zone of voluntary stable [7]

Where:

MH: Hip flexion moment

MH': Hip extension moment

P is the reaction force from heel

P' is the reaction force from toe.

Hip flexion moment applied. So, when we apply the hip flexion moment, it moves the load line.

Figure 20a: Heel contact

When we apply the hip flexion from heel, the region behind that load line QP is the zone of stability. As long as the knee is located behind the load line QP (pic a), It is stable.

Figure 20c: Toe off

Hip extension moment here applied. when we apply the hip extension moment and move the load line behind the knee. As long as the knee is in front of anywhere in this region in front of this Q'P' (pic c), it can initiate flexion moment.

Figure 20b: Superimposed (zone of voluntary stability)

We superimpose these two stages in this cross-hatched region where the two meet is known as the zone of voluntary stability. That means the stability region of the heel and toe overlaps. If the ICR is located in this cross-hatched, with this hip moment in this voluntary zone, we can control stability at heel strike and initiation of flexion of the swing. For prosthetic patients, this zone shrinks, as shown in **figure 21.b**, because of the number of moments of the hip thrust, they would be able to apply lower than an able-bodied person. So, this zone is smaller than the zone in a person who has normal hip strength.

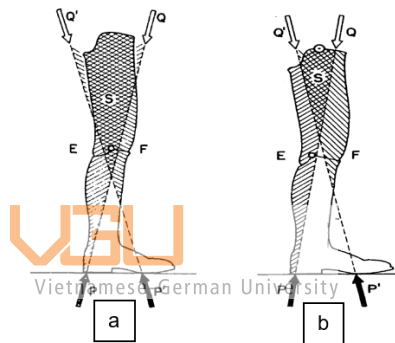


Figure 21 Normal human leg is zone of stability compare to amputee human leg is zone of stability [7]

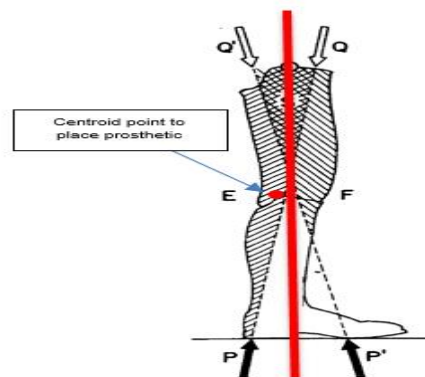


Figure 22 Represent the point place prosthetic [7]

Applied to our design as followed:

The hip anchor line is the line that is perpendicular to the ground from the hip. From this line, the alignment of the knee will be put such that the centroid of the prosthetic knee is located 9 to 15-millimeter posterior to the centroid line, as shown in figure 22.

The reason to do that is since at heel contact, we are ensuring even with minimal hip strength, the centroid will remain behind the load line. Offset it from where the normal knee would be to ensure stability at heel contact. When the knee is hyper-stable at heel contact, the center of gravity is further behind where it should normally be. It will provide a flexion moment to initiate a heel strike for the swing.

Point 2: Ability to Increase the ability user use for walking on uneven terrain (Toe clearance)

Solution: Maximized toe clearance during swing. (2nd)

This is the clearance of the toe from the ground during the swing, starting from the angle of 23° of the hip flexion cycle and 49° of the knee flexion cycle [9]. Normal walking needs a small amount of toe clearance. But for more far purposes than walking, the toe clearance index must be increased. The lowest toe clearance for a prosthesis user may occur at numerous combinations of hip and knee flexion. The greater the minimum toe clearance, the less the chances of tripping during the swing phase. The toe clearance can represent through the Instantaneous center height range (Q) because the AC can assume as the ground link and connect to the shank, so only the BD moves with the thigh. The hip movement so can consider the height of the highest point of moving side a 4 bar mechanism (D) or (B) up to the type of linkage, and the ICR (P) is a toe clearance range of the design [10], present in **figure 23** (with an example of Crank Quadrilateral Linkage).

Applied to our design

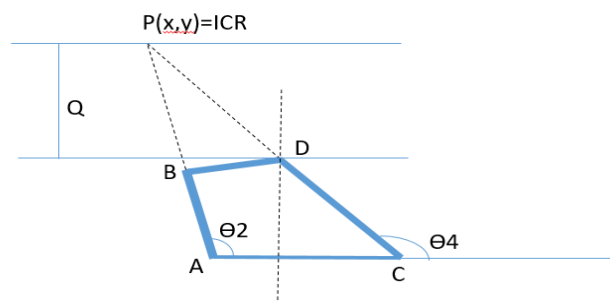


Figure 23 Represent the ICR (Instant Centre of Rotation)

Figure 23 represents the ICR and the Q (the distance from the highest point of 4 bar mechanism D to the ICR called the Instantaneous center height range). To prevent hyper-stabilization, the height limit of the Instantaneous center height range (Q) maximum is constrained to 400mm, and the minimum height is set at 100mm. [10]

So Q height constraint is $100 < Q < 400$ mm [10]

Point 3: The design material and price point (Price point)

Solution: Optimize the material selection (1st)

To make the design reduce weight and cost compared to other previous designs. We study the human leg based on the weight line to find out the tensile strength. We take a regular person's data with a mass of 750 N=75kg as well as the Tensile Force= 2050N, With the average Young's modulus of normal male bone to be 17.0×10^9 Pa The tibia metrics 0.4 m in length, 1.3 cm in average inner diameter, and 2.5 cm in average outer diameter. To replicate the prosthesis limit weight we have the individual stand in 1 foot, the person weighs 750N but because the person is on their toe, tensile forces in the calf are required to stretch the toe downward and lift them off the ground (generating a moment at the ankle joint) (creating a moment about the ankle joint) Given that the person is in static equilibrium, the tibia must react to these forces in a way that ensures that the sum of all forces acting in the y-direction (up and down the leg) is zero.

tensile force: $\vec{F}_{tensile}$

$$\vec{F}_{tensile} + \vec{W} + \vec{F}_{Tibia} = 0 \tag{3.3}$$

$$2050 + 750 + \vec{F}_{Tibia} = 0$$

$$\vec{F}_{Tibia} = 2800N$$

=> so the tibia is experiencing a total force in the y-direction of 2800N

Figure 24 demonstrate the free body diagram of the tibia under the mass

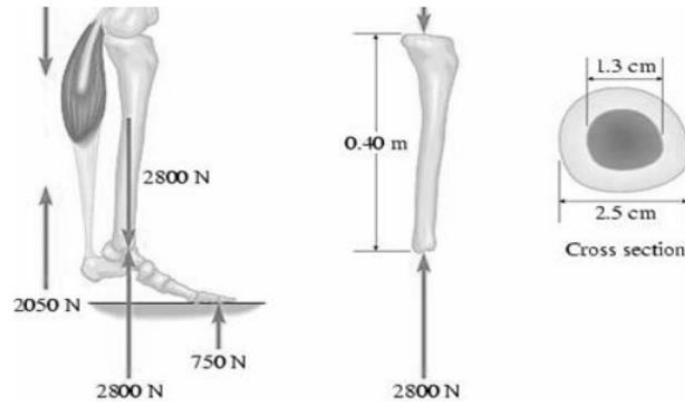


Figure 24 Free body diagram of the leg

Area of a circle however the trick here is that the marrow won't take any compressive forces. The actual structural x-sectional area is only contributed by the bone-bit

$$A = \pi(r_{outside}^2 - r_{inside}^2) = 3.58 \times 10^{-4} m^2 \quad (3.4)$$

Stress is force/area

$$\sigma = \frac{F_{tibia}}{A} = \frac{2800N}{3.58 \times 10^{-4}} = 7.82 Mpa \quad (3.5)$$

Stain in the x and z directions are clearly 0, so we don't need Poisson's ratio, ν , and E is the modulus of the bone

Hooke's Law (in y-direction only)

$$\epsilon_y = \frac{1}{E}(\sigma_y - \nu(\sigma_z + \sigma_z)) = \frac{1}{17.0 \times 10^9} (7.82 \times 10^3) = 4.6 \times 10^{-7} pa = 0.46 MPa \quad (3.6)$$

To make sure the foot suitable for everyday activities like walking in high speed or up and down the stair, the foot must have to please the request of applied 5 times the body weight

That mean the stress and stain will equal to 5 time this number

$$\sigma = \frac{F_{tibia}}{A} = 7.82 Mpa \times 5 = 39.1 Mpa$$

$$\epsilon_y = \frac{1}{E}(\sigma_y - \nu(\sigma_z + \sigma_z)) = 0.46 MPa \times 5 = 2.3 MPa$$

From the tensile strength σ_{tibia} we can minimize the material based on the previous we see the most common material is stainless steel, titanium, Aluminum

We compare the material

Article No.	Material	Tensile Strength (Mpa)	Density	Price per flat sheet per kg
316L	Stainless steel	175	7980 kg/m ³	3.8usd
Ti GR.	titanium	880	4540 kg/m ³	38usd
304	Stainless steel	210	8000 Kg/m ³	3.2usd
1100	Al	76Mpa	2710kg/m ³	2.36usd

Table 2 Material comparison table

Applied to our design

After comparing based on *Tensile Strength* (39.1Mpa) we see that stainless steel has a much more requirement tensile strength with a much higher price than necessary, so another material has to be considered. AL is the material with the nearest tensile strength requirement. So we try to take the material from the AL and calculate the required safety factor.

$$\text{Safety factor: } SF = \frac{\sigma_{yield}}{\sigma_{totmax}} = \frac{\sigma_{tibia}}{\sigma_{totmax}} \quad (3.7)$$

$$\sigma_{totmax} = \frac{\sigma_{tibia}}{SF} = \frac{39.1Mpa}{2} = 19.55Mpa \quad (3.8)$$

In here we take the safety factor is 2 approximate for human bones

So base on the table 2 and the safety factor we see the AL is the most suitable and optimize material for the 4 bar part.

Chapter summary

This chapter provides the conceptual design, a study of the basic condition of the 4-bar polymeric knee design, and what is needed to improve the design to make the new design different from the previous benchmark design based on the focus point of the 4-bar prosthetic knee.

From the study of 4 bar and human knee mechanics, the design must have the following condition to fully function as a knee replacement prosthetic:

- Design use Binary link
- Design is a Crank Rocker
- Design must have 1 degree of freedom
- Design knee must be Non-Grashof ($S+L>P+Q$)

The new design criterion following condition must be:

- Design centroid of prosthetic knee is located 10 to 15-millimeter posterior.
- Design constrain the Instantaneous center height range Q to $100 < Q < 400$ mm
- Design material is focus is aluminum

Chapter 4: Parametric Design

Introduction

Following the concept to design and study of previous benchmark design, the parametric design is the next step need to conduct. To get the design parameters, we used an available visual design model of four bar 3R20 prosthetic knee to optimize with the corresponding edge parameters (**table 3**).

L1=27.5mm	L3=22mm
L2=34mm	L4=44.5mm

Table 3 Previous parametric

Design configuration

The visual design only provides a range of motion. However, it does not provide any support system or detail to function. Based on the measurements and range of motion we compared the conditions and found this model is suitable for applying the optimization. To match the optimization, first, we compare the scale and raise the parametric scale of the old model to suit the design. The material of 3R20 is 316 stainless steel with a Tensile Strength of 175Mpa, which means two times stronger than the Aluminum Strength 76Mpa. In order to optimize the design, we have to scale the parameter also.

The length of base prosthetic knee links (**table 4**) is chosen:

L1=55mm	L3=44mm
L2=68mm	L4=89mm

Table 4 New base parametric

After chosen the parametric we have to make sure it satisfied the condition constrain

Condition Constrain:

- Design use Binary link
- Design is a Crank Rocker

- Design must have 1 degree of freedom
- Design knee must be Non-Grashof ($S+L>P+Q$)
- Design centroid of prosthetic knee is located 10 to 15-millimeter posterior.
- Design constrain the Instantaneous center height range Q to $100 < Q < 400$ mm
- Design material suitable

Number of links = 4

Number of 1 DOF joint = 4 (4 Revolute) $\Rightarrow M=1$ (**the design satisfy DOF condition**)

The knee is must defined as Crank Rocker, so in this part we need to defined what kind of crank rocker design is suitable for our condition limitation Q range to $100 < Q < 400$ mm to this we chose **Quadrilateral Linkage (RRRR)** show in figure.

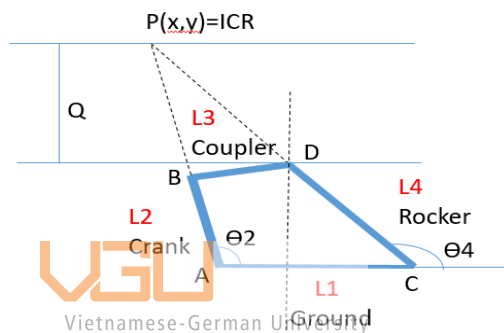


Figure 25 Represent the ICR and Quadrilateral linkage

Which mean the motion part is occurred and limited by the coupler. This is achieving by Using AutoCAD software to draw a structure diagram and consider each position to infer the parameter show in Figure.

The result shows the $Q=144.606$ mm **the design satisfies constrain the Instantaneous center height range Q to $100 < Q < 400$ mm**

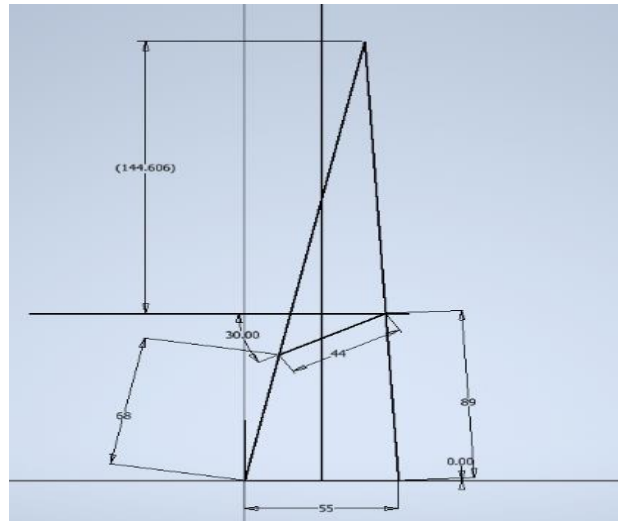


Figure 26 Represent the 4 bar on inventor concept

Quadrilateral Linkage (RRRR) include:

Ground Link Length (Link 1)	l1=55mm=G
Crank Length (Link 2)	l2=68mm=In
Coupler Length (Link 3)	l3=44mm=Coupler
Rocker Length (Link 4)	l4=89mm=Out

Table 5 New design base parametric

After assign the parameter to the linkage system we have to compare again with the condition to see if it satisfy the condition or not.

Non-Grashof condition:

S=l3=44mm=Coupler	S+L=133mm	S+L>P+Q 133>123 Non-Grashof
L=l4=89mm=Out (rocker)		
P=l2=68mm=In (Crank)	P+Q=123mm	
Q=l1=55mm= G		

The design is non-Grashof → Correct to requirement of knee

After having the basic base parameter, we design the parameter for each part to be suitable for the criterion.

The design centroid of the prosthetic knee is located 9 to 15-millimeter posterior. To solve this, we chose the Coupler L3 because the L3 is connected directly to the body part and directly affects the centroid line. The design uses the Trapezium geometry represented in **figure 27**. Using this geometry, we can easily calculate the centroid of the Trapezium, in this case, can be considered the centroid of the prosthetic (remaining part is a simple binary link).

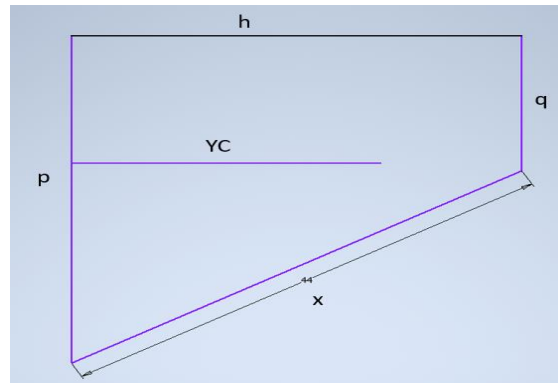


Figure 27 Represent of design L3 coupler on inventor



Vietnamese-German University

Input condition:

- $X=44\text{mm}$ (the length of coupler)
- $15.746\text{mm} < Y_c < 21.696\text{mm}$ (distance from centroid to load line located 9-15 millimeter posterior in y-direction)
- $q > 15\text{mm}$ (due to the limitation of bearing available)
- The angle of p and x is 60°

We got the centroid of trapezoid formula

$$C = [h/2, \frac{(q+2p)h}{3(p+q)}] \quad (4.1)$$

We got the equation of q flowing

$$\sin \alpha = \frac{h}{x} \quad (4.2)$$

$$\sin 60 = \frac{h}{44} \Rightarrow h = 44 \cdot \sin 60$$

$$p = q + p'$$

$$p = q + \sqrt{44^2 - h^2} \text{ (Pythagorean)}$$

Replace equation we got

$$y_c = \frac{[(q + 2(q + \sqrt{44^2 - h^2})) h]}{3[(q + \sqrt{44^2 - h^2}) + q]} \quad (4.3)$$

After finding the equation flowing q we replace q from 15 to infinity using excel we got the table of Yc according to q(mm).

q (mm)	p (mm)	Yc (mm)
15	37.0002	21.73941
15.04	37.0402	21.73528
15.08	37.0802	21.73116
15.12	37.1202	21.72706
15.16	37.1602	21.72297
15.2	37.2002	21.71889
15.24	37.2402	21.71483
15.28	37.2802	21.71078
15.32	37.3202	21.70674
15.36	37.3602	21.70271
15.4	37.4002	21.69869
15.44	37.4402	21.69469
15.48	37.4802	21.6907
15.5	37.5002	21.68871

Table 6 Result of p and Yc according to q

We compare the condition of $Y_c 15.746\text{mm} < Y_c < 21.696\text{mm}$ and the limit of q. Using the resulting number to draw a trend forecast **figure 28**, we can see that the higher the q, the lower the Yc. So to optimize the material, the index q must be close to the limit, which is 15. As we can see from the **table 6**, with q=15.4, we have Yc=21.698, which is close to the upper limit of Yc and close to the minimum limit of q. The smaller q means less material consumption and less expensive the material is to manufacture. So the value of q=15.4 is selected for the design.

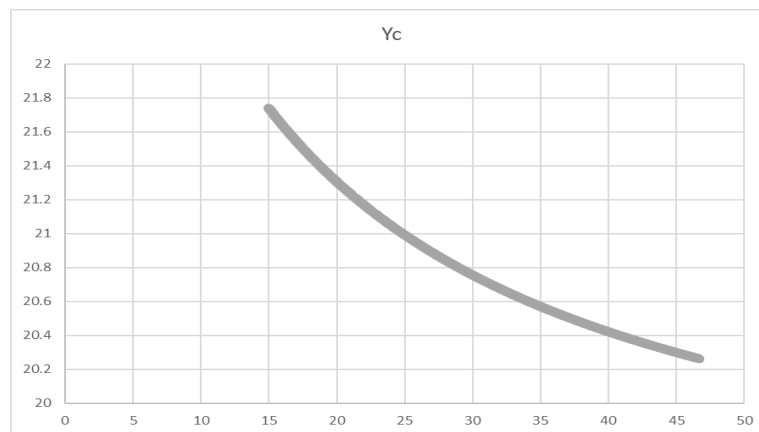


Figure 28 Trend forecast of Yc according to q

After having q we easily having p and h . Finally, we have the design with trapezoid parameter of $p=15.418\text{mm}$; $q=37.418\text{mm}$; $h=38.105\text{mm}$ show in figure x

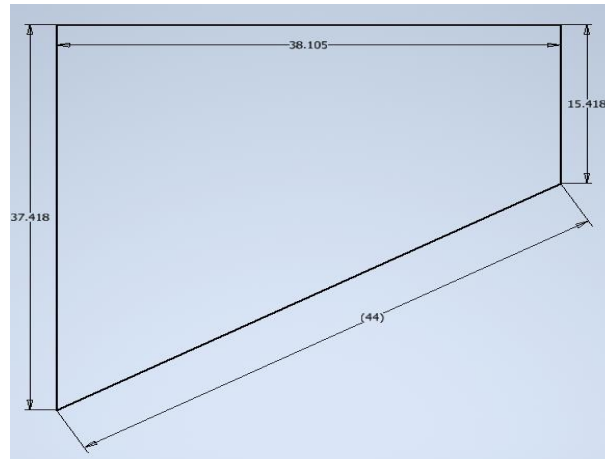


Figure 29 Parametric design of L3

The coupler design is the main major change in the design. The remaining part is the binary link with the **table 5** parameter.

Shaft radius calculation and material selection

After having the design parametric, we continue to calculate the shaft.

To connect each part, we use a shaft and bearing, which is connected to a bearing attached to the body part.

Calculation of shaft diameter when lifting load, **figure 30** is a free body diagram of the shaft when force reacting.

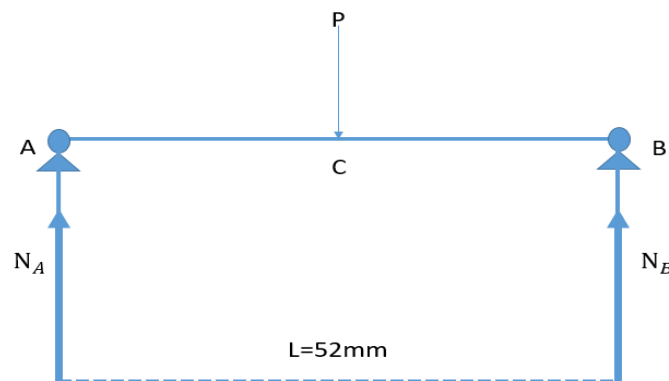


Figure 30 Force distribution free body diagram on the shaft axis

With the input parameter load at least 75kg person we estimate the force estimate the force P in the diagram over about 750N.

Calculating the bonding reaction at the bearing inner shaft:

$$\begin{cases} \sum \mathbf{F} = \mathbf{0} \\ \sum \mathbf{M} \text{ at } A = \mathbf{0} \end{cases} \Leftrightarrow \begin{cases} N_A + N_B - P = 0 \\ P * AC - N_B * AB = 0 \end{cases} \quad (4.4)$$

$$\begin{cases} N_A + N_B = P = 750\text{N} \\ 2700 * AC - N_B * AB = 0 \end{cases} \Leftrightarrow \begin{cases} N_A + N_B = P = 750\text{N} \\ 750 * 26 - N_B * 52 = 0 \end{cases}$$

$$\begin{cases} N_A = 375\text{N} \\ N_B = 375\text{N} \end{cases}$$

Draw the internal force diagram of the shaft including the diagram of shear force and bending moment

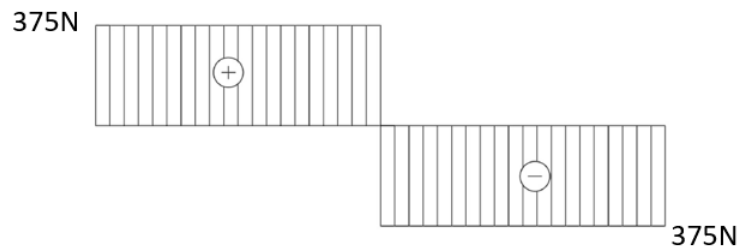


Figure 31 Shaft shear force diagram



Figure 32 Shaft bending moment diagram

Static strength test shaft:

To prevent excessive plastic deformation or failure due to sudden overload, the shaft should be tested for static strength. According to [1], page 200, the test formula

The test has the form:

$$\sigma = \sqrt{\sigma^2 + 3\tau^2} \leq \sigma \quad (4.5)$$

$$\sigma = M_{Max} / (0.1d^3) = \frac{9750}{0.1 \cdot 10^3} = 97.5 \text{ N/mm}^2 \quad (4.6)$$

$$\tau = T_{Max} / (0.2d^3) = \frac{375}{0.2 \cdot 10^3} = 1.87 \text{ N/mm}^2 \quad (4.7)$$

$$\sigma = \sqrt{\sigma^2 + 3\tau^2} = \sqrt{97.5^2 + 3(1.87)^2} = 97.5 \text{ Mpa} \quad (4.8)$$

With the same calculation to the inner shaft the outer shaft with the length of 85mm have the result,

$$\sigma = \sqrt{\sigma^2 + 3\tau^2} = \sqrt{159.37^2 + 3(1.87)^2} = 159.4 \text{ Mpa}$$

With the first selection we select the Aluminum material

$$[\sigma] = 0.8 * \sigma_{ts} = 0.8 * 76 = 60.8 \text{ Mpa}$$

$$\sigma_{inner} == 97.5 \text{ Mpa} \geq 60.8 \text{ Mpa}$$

$$\sigma_{outer} == 159.4 \text{ Mpa} \geq 60.8 \text{ Mpa}$$

So the material is not suitable for shaft cause the shaft affect by force applied of the whole structure the material with the more tensile strength is selected

304 stainless steel with $\sigma_{ch} = 210 \text{ Mpa}$ is selected

$$[\sigma] = 0.8 * \sigma_{ts} = 0.8 * 210 = 168 \text{ Mpa} \quad (4.9)$$

$$\sigma_{inner} == 97.5 \text{ Mpa} \leq [\sigma] = 168 \text{ Mpa}$$

$$\sigma_{outer} == 159.4 \text{ Mpa} \leq [\sigma] = 168 \text{ Mpa}$$

So the given inner shaft is suitable enough so both inner and outer shaft is suitable with the d of 10mm

Shank pine diameter calculation

To ensure the safety of the design we calculate the shank pipe diameter

Calculation for the shank pipe to for Inner D=35mm

Cylinder intersection area: R of outer diameter = 40mm

$$A = \pi \cdot r_{solid}^2 = \pi \cdot (R^2 - r^2) \quad (4.10)$$

Stress formula:

$$\sigma = F/A \quad (4.11)$$

Since

$$F = 2800 \text{ N (the maximum weight acted on one leg)}$$

Tensile strength of aluminum = 90 MPA = $9 \times 10^7 \text{ N/m}^2$

With

$$A = \pi \cdot r_{solid}^2 = 3.11 \times 10 \times 10^{-5} = \pi \cdot (0.04^2 - r^2)$$

$$\sigma = F/A$$

$$\Rightarrow A = F/\sigma = 2800/(9 \times 10^7)$$

$$A = 3.11 \times 10 \times 10^{-5} (\text{m}^2)$$

$$\Rightarrow r=0.039 \text{ m} = 39.8 \text{ mm}$$

Therefore, we can assure that the shank pipe can handle 2800N at long as the ID is smaller than 39mm

With the result above, we decide to Round down the inner diameter to 35 mm for safety assurance.

Simulation

Position analysis

To estimate the motion of the 4-bar, we use Matlab position analysis based on the Loop Closure Method and New-ton Raphson method. Using position analysis, we will find out the orientation of each link relative to the fixed link for a given configuration of four bar mechanism. The length of the links and the angle made by the link to that is the angle theta two are the inputs needed for the position analysis the angles theta 3 and theta 4 are the outputs. This method proceeds to find out the ICR as well as the displacement of point to compare gait cycle simulation with knee angle in the gait cycle to make sure the maximum knee flexion and angle of load line for stability.

Matlab simulate

Input: $L_1, L_2, L_3, L_4, \theta_2$

Output: θ_3, θ_4

Is show as **figure 28** as a 4 bar mechanism in MathLab

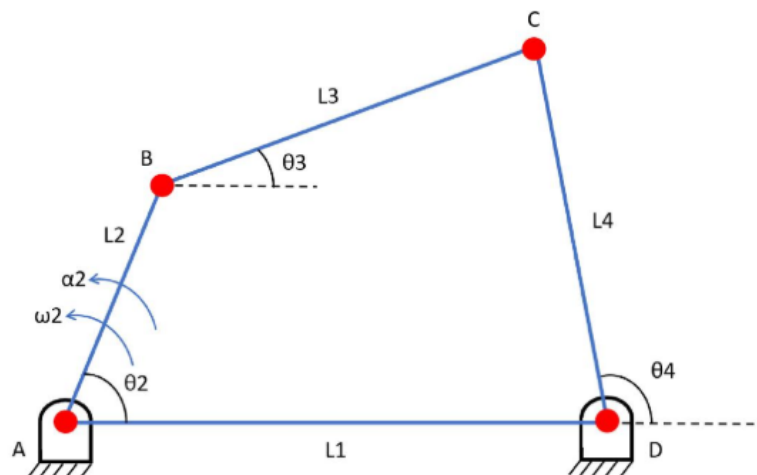


Figure 33 Represent the 4 bar on Matlab simulation

we will use loop closure method for position analysis

Starting from point D the length of each link expressed in horizontal and vertical component in term of i and j

Loop Closure Method

$$\begin{aligned}
 & -l_1 \hat{i} \\
 & l_2 \cos \theta_2 \hat{i} + l_2 \sin \theta_2 \hat{j} \\
 & l_3 \cos \theta_3 \hat{i} + l_3 \sin \theta_3 \hat{j} \\
 & l_4 \cos (180^\circ + \theta_4) \hat{i} + l_4 \sin (180^\circ + \theta_4) \hat{j}
 \end{aligned}$$

Using loop closure method for A, B, C, D sum of component is 0

So the I component and J are also 0

\hat{i} component

$$f_1 = -L_1 + L_2 \cos \theta_2 + L_3 \cos \theta_3 + L_4 \cos (180 + \theta_4) = 0$$

\hat{j} component

$$f_2 = L_2 \sin \theta_2 + L_3 \sin \theta_3 + L_4 \sin (180 + \theta_4) = 0$$

So we have 2 equations with 2 unknown

But these equation is non-linear and cannot be solve directly, We using the New-ton Rasphson method to sold the equation

Newton-Raphson method

$$x_{t+1} = x_n - \frac{f(x_n)}{f'(x_n)} \tag{4.12}$$

$$f'(x_n) \Delta x = -f(x_n) \tag{4.13}$$

the Newton-Raphson method is iterative Take the initial guess for the unknown variable this is called the initial solution then the method keeps on doing iterations till the solution converges here is the formula used in the Newton-Raphson method to calculate the values of unknown variables in each iteration

$$\begin{bmatrix} \frac{\partial f_1(x_1, x_2)}{\partial x_1} & \frac{\partial f_1(x_1, x_2)}{\partial x_2} \\ \frac{\partial f_2(x_1, x_2)}{\partial x_1} & \frac{\partial f_2(x_1, x_2)}{\partial x_2} \end{bmatrix} \begin{Bmatrix} \Delta x_1 \\ \Delta x_2 \end{Bmatrix} = - \begin{Bmatrix} f_1(x_1, x_2) \\ f_2(x_1, x_2) \end{Bmatrix} \tag{4.14}$$

delta x is the difference of value x in each iteration in matrix form the Newton-Raphson method gives this equation

here is the matrix form of Newton-Raphson method in terms of theta 3 and theta 4.

$$\begin{bmatrix} \frac{\partial f_1}{\partial \theta_3} & \frac{\partial f_1}{\partial \theta_4} \\ \frac{\partial f_2}{\partial \theta_3} & \frac{\partial f_2}{\partial \theta_4} \end{bmatrix} \begin{Bmatrix} \Delta \theta_3 \\ \Delta \theta_4 \end{Bmatrix} = - \begin{Bmatrix} f_1(\theta_3, \theta_4) \\ f_2(\theta_3, \theta_4) \end{Bmatrix}$$

$$\theta_3^{n+1} = \theta_3^n + \Delta \theta_3$$

$$\theta_4^{n+1} = \theta_4^n + \Delta \theta_4$$

the derivatives of functions f_1 and f_2 are substituted in this equation

$$\begin{bmatrix} -l_3 \sin \theta_3 & -l_4 \sin(180 + \theta_4) \\ l_3 \sin \theta_3 & l_4 \sin(180 + \theta_4) \end{bmatrix} \begin{Bmatrix} \Delta \theta_3 \\ \Delta \theta_4 \end{Bmatrix} = - \begin{Bmatrix} -l_1 + l_2 \cos \theta_2 + l_3 \cos \theta_3 \\ + l_4 \cos(180 + \theta_4) \\ l_2 \sin \theta_2 + l_3 \sin \theta_3 \\ + l_4 \sin(180 + \theta_4) \end{Bmatrix}$$

How to solve the angle of θ_3, θ_4 based on Input: $L_1, L_2, L_3, L_4, \theta_2$, Output: θ_3, θ_4

Step 1: Input the known value $L_1, L_2, L_3, L_4, \theta_2$

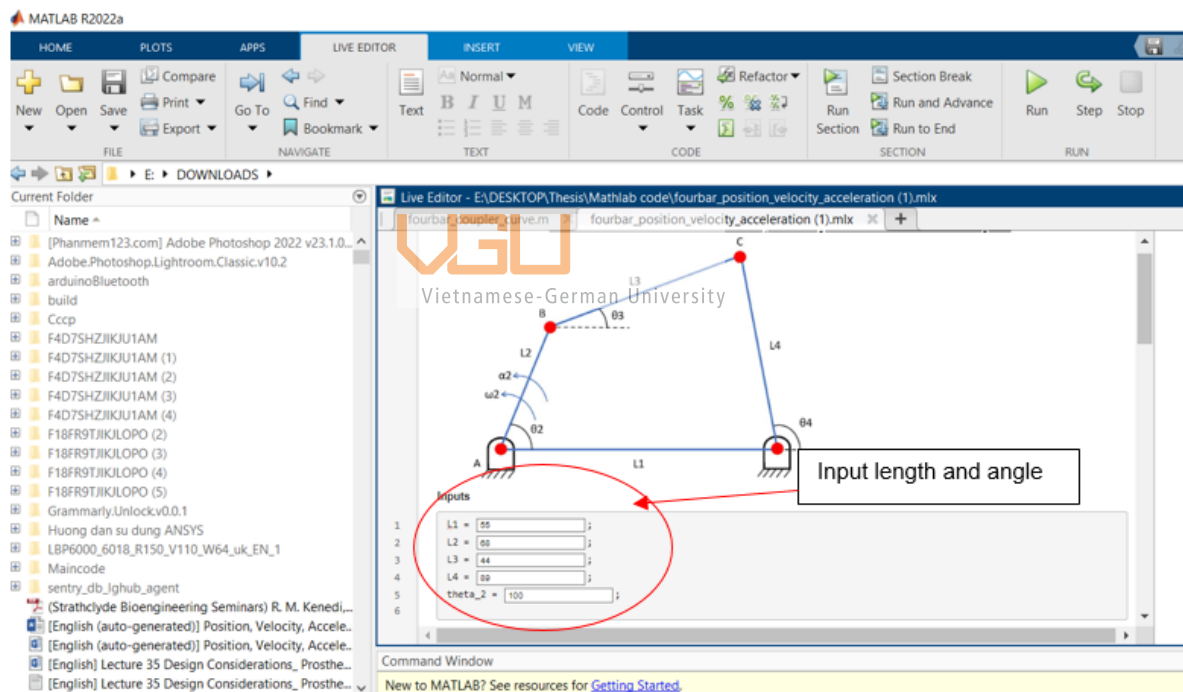


Figure 34 Step 1 of Matlab simulation

Step 2: input initial guess for the unknown variable

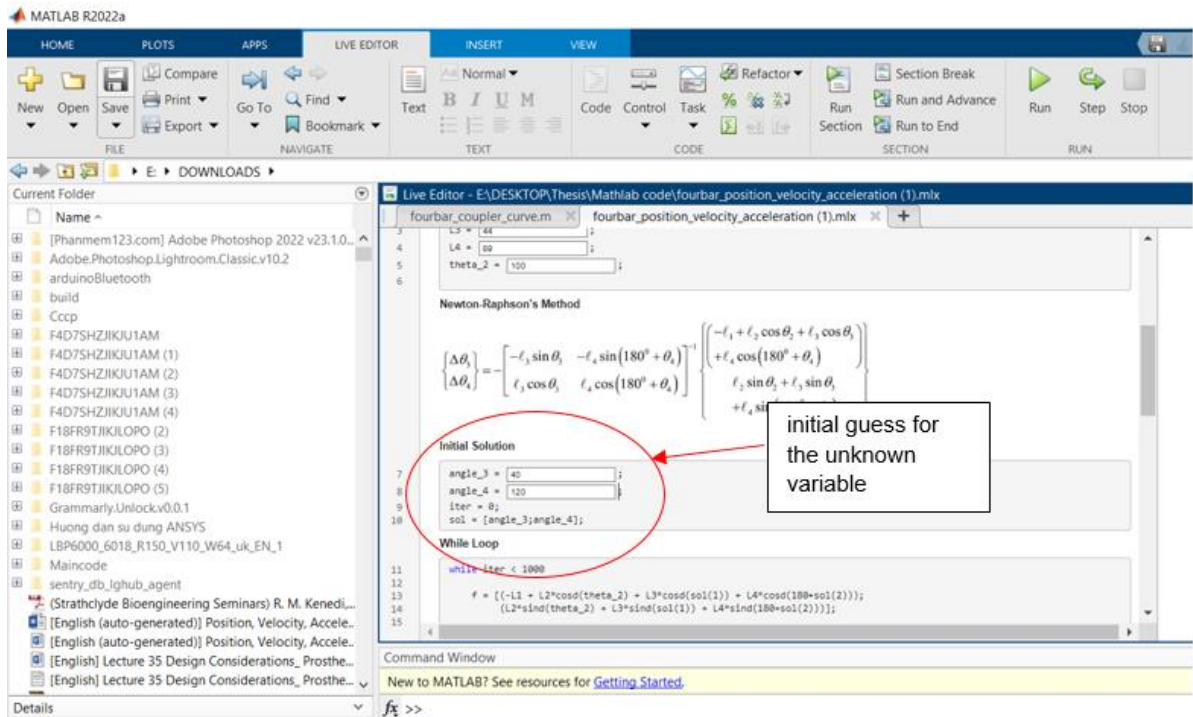


Figure 35 Step 2 of Matlab simulation

Step 3: Using While loop to solve then the method keeps on doing iterations till the solution converges

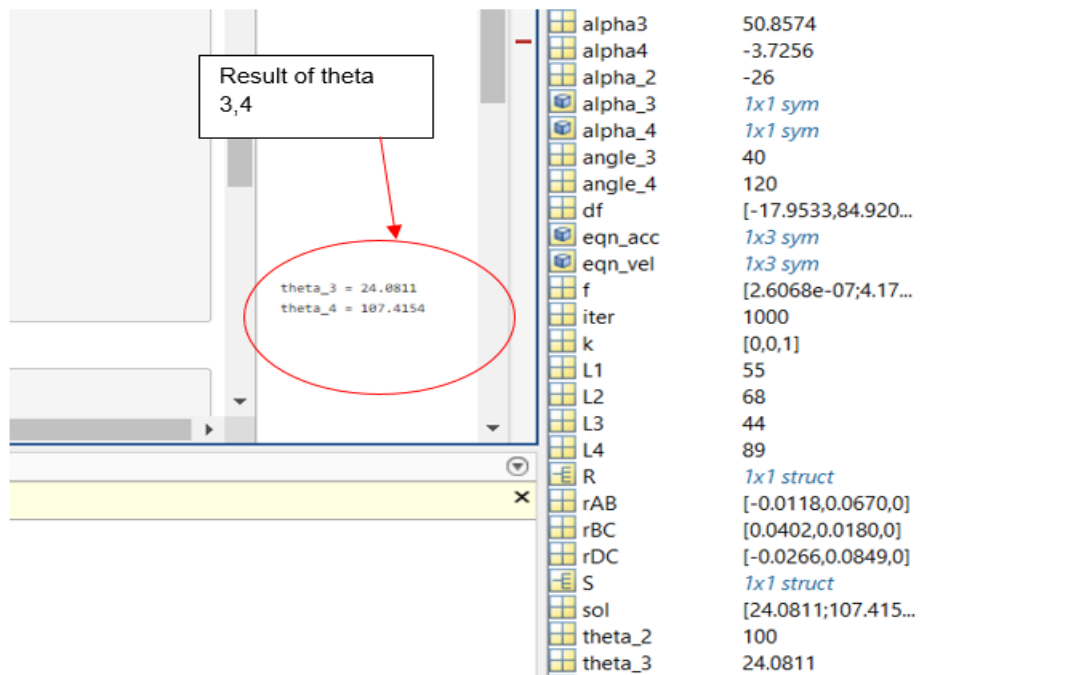


Figure 36 Step 3 of Matlab simulation

Matlab code to estimate theta 3,4

```

while iter < 1000
    f = [(-L1 + L2*cosd(theta_2) + L3*cosd(sol(1)) + L4*cosd(180+sol(2)));
        (L2*sind(theta_2) + L3*sind(sol(1)) + L4*sind(180+sol(2)))];

    df = [(-L3*sind(sol(1))), (-L4*sind(180+sol(2))) ;
        (L3*cosd(sol(1))), (L4*cosd(180+sol(2)))];

    sol = sol - (df\f);
    iter = iter + 1;
end
theta_3 = sol(1)
theta_4 = sol(2)

```

After finding theta 3,4 we use the result to find the instant center of rotation.

The instant center of rotation (ICR) between the femur and the shank can be considered the intersection of two straight lines this show in **figure 37** as an example.

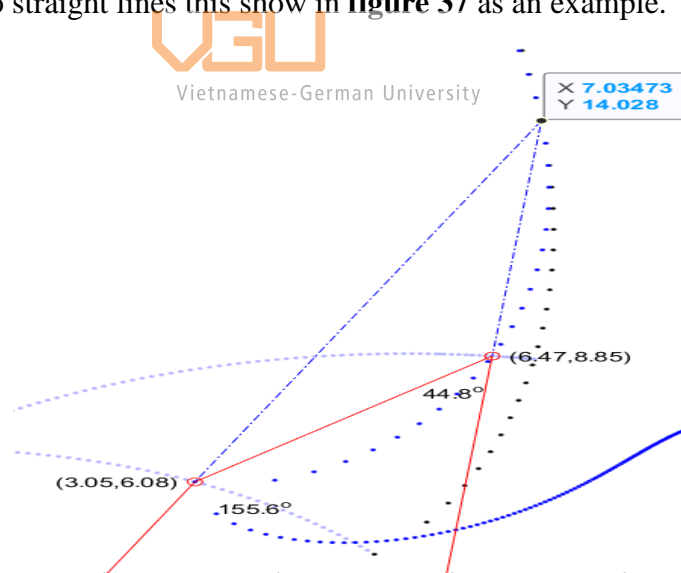


Figure 37 Represent the 4 bar ICR trajectory on Matlab simulation

The first line passes through AB, while the second line passes through CD.

The ICR coordinates can be obtained from the following equation as follow:

Enlengthen the input (crank) AB and output (rocker) CD link to where it intersects, then determine the coordinates of the point of intersection. By using this method in Mathlab, the

coordinate of the ICR is found after simulation to get the ICR and angle data. We use this to see the gait cycle and the specification of the design.

Result: Compare gait cycle for prosthetic after simulation

Gait cycle:

To find maximum knee flexion and angle of load line for stability, the gait cycle when swing phase is taken into comparison. There are five distinct events that happen in a walking cycle stance phase: thigh extension, toe-off, maximum knee flexion, maximum thigh flexion, and heel contact, illustrated in **figure 38**. According to experimental data from numerous studies, the swing phase in the / polycentric knee was only taken into consideration when comparing corresponded to between knee flexion.

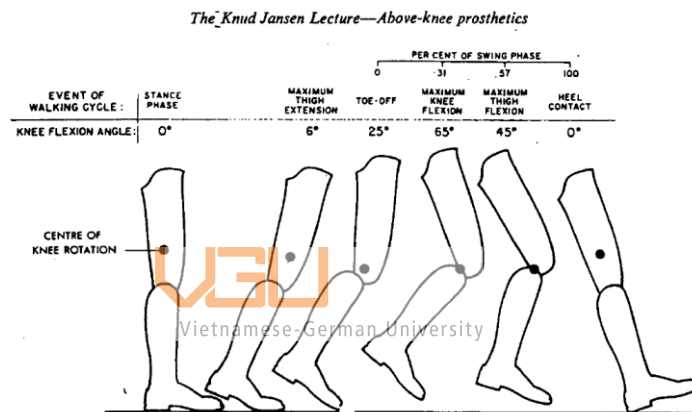


Figure 38 Represent the gait cycle of a normal person

Gait Cycle % Swing phase	Knee Angle	Swing phase segmentation	The orientation of links AB and CD with respect to AC [Degree] represent in figure 25		ICR coordinates [mm]	
			θ_2	θ_4	x	y
0%	25	Toe-off	66.4	94.8	6.16355	15.4109
31%	65	Maximum knee flexion	47.9	81.5	3.51354	7.54795
57%	45	Maximum thigh flexion	56.3	81.2	5.49078	10.6306
100%	0	Heel contact	83.5	95.6	2.97128	25.9832

Table 4 Result of gait cycle angle compare to 4-bar mechanism after simulate

The result of the comparison shows the corresponding angle according to the real angle and maximum knee and thigh flexion. This number provides us with the maximum angle of the design, and the latter will be used to compare with the real design prototype.

The result taken from Matlab simulation represents **Figures 39,40,41,42**. The 4 bar ICR trajectory result on Matlab simulation corresponding with the gait cycle.

Class 3 Rocker-Rocker-Rocker Non-Grashof Mechanism

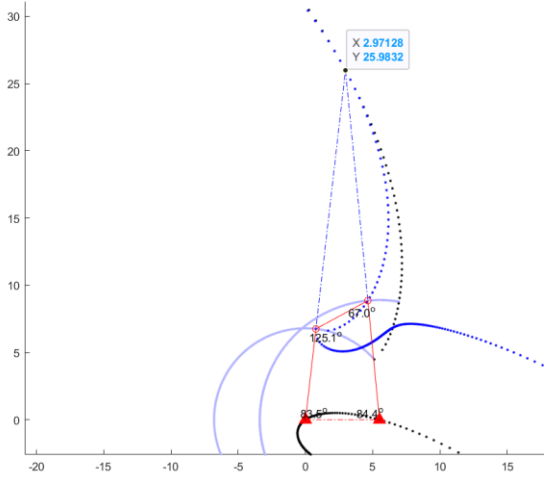


Figure 39 Angle and ICR of heel contact

Class 3 Rocker-Rocker-Rocker Non-Grashof Mechanism

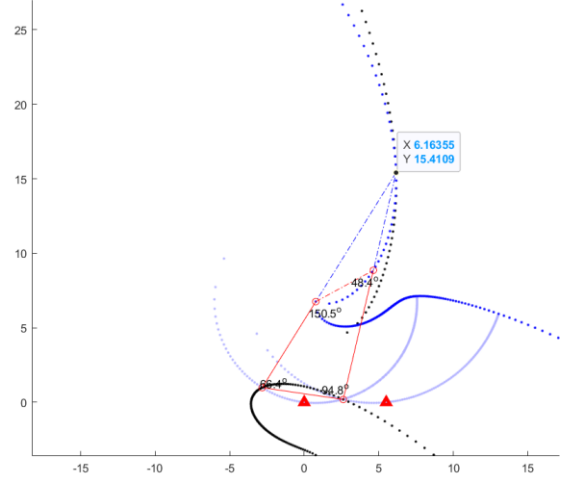


Figure 40 Angle and ICR of toe off

Class 3 Rocker-Rocker-Rocker Non-Grashof Mechanism

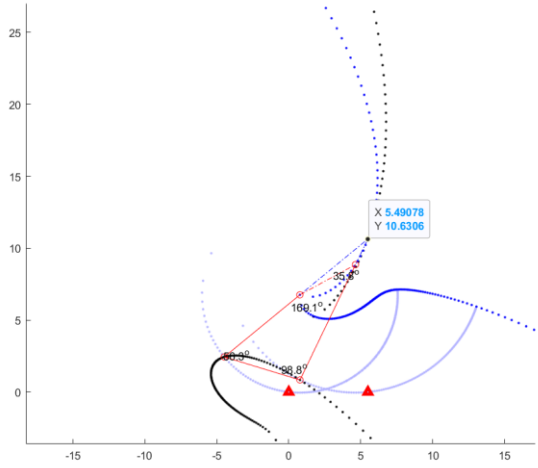


Figure 41 Angle and ICR of maximum thigh flexion

Class 3 Rocker-Rocker-Rocker Non-Grashof Mechanism

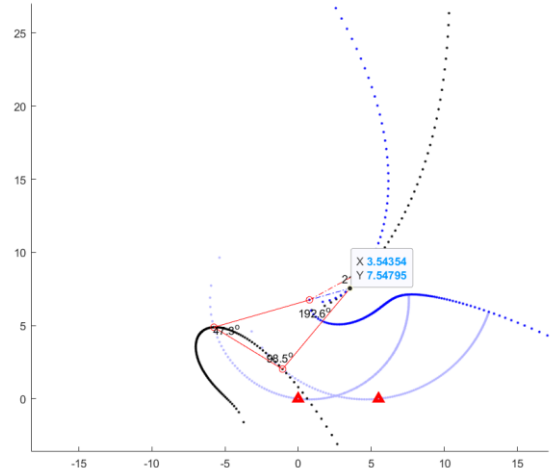


Figure 42 Angle and ICR of maximum knee flexion

Chapter 5: Detail design

Introduction

When having the parametric, we begin to compute an actual concept of 4 bar knee. We use the parametric also learning from the previous design to compute the CAD design of a 4-bar system. This CAD will be simulated using FEA to simulate the static stress, displacement, also safety factor of the design.

Prosthetic leg Structural Diagram

A System of design includes two major main parts: the 4-bar and a shank connected through a coupling. The 4 bar knee consist of 4 part: L1, L2 (crank), L3 (coupling), L4 (rocker) with screw holes on parts L1 and L3 to connect with other compartments:

- A pyramid adapter on top of the 4 bar for connecting to the residual limb socket
- A pipe Coupling under the 4 bar for connecting to the shank pipe

The shank shaft will act as a shank that connects from the 4 bar mechanism to the foot base, made up of the length of the leg shown as **figure 43**.

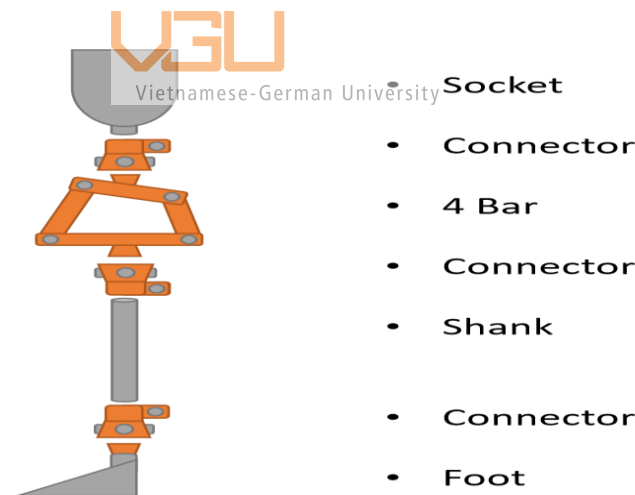


Figure 43 Represent Prosthetic Leg Structural

The 4-bar mechanism is divided into two main parts: the 4-bar body components: L1, L2 (crank), L3 (coupling), L4 (rocker), and the shaft. The shaft is connected to the body with an attached bearing that uses for rotating motion. A force applied on a pyramid adapter will

affect the 4-bar body. The same adapter is attached to L1 and connects the prosthetic leg with the amputee's body mass.

Manufacture part

Part L1

Part L1 shown in **figure 44** serves as the base for the 4 bar integrated with a slot that limits the trajectory of the leg to 90 degrees, creating comfort for the patient when sitting. With 4 M5 screw holes below in order to connect with the Shank Coupling. **Figure 44** shows the L1 part in the inventor show and **figure 45** shows L1 drawing.

Material: Aluminum



Figure 44 Part L1 in Inventor

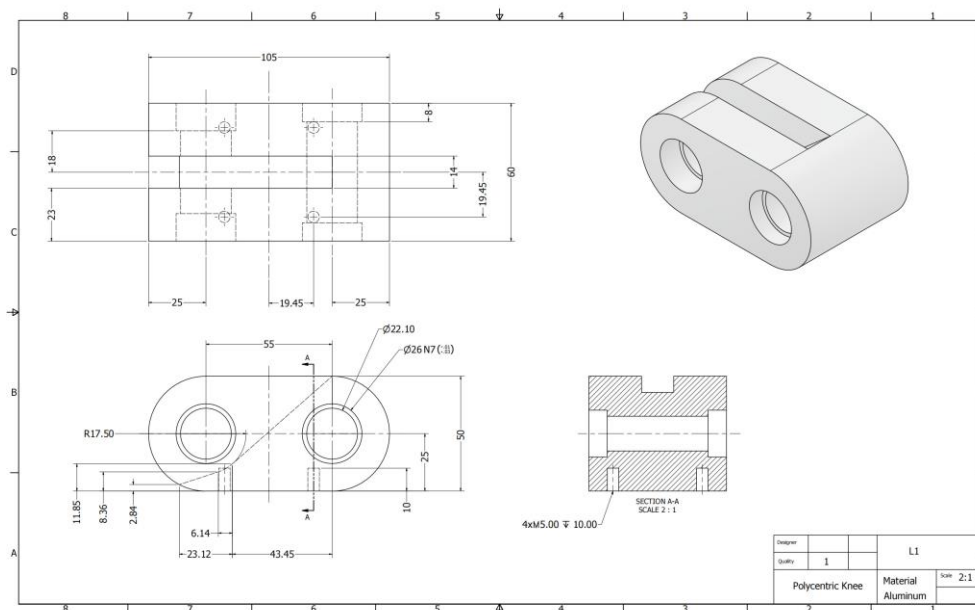


Figure 45 Part L1 Drawing

Part L2

L2 serves as Crank Link for the 4-bar mechanism shown in **figure 46** connecting L1 with L3 **Figure 46** shows the L2 part in the inventor show and **figure 47** shows L2 drawing.

Material: Aluminum



Figure 46 Part L2 in Inventor

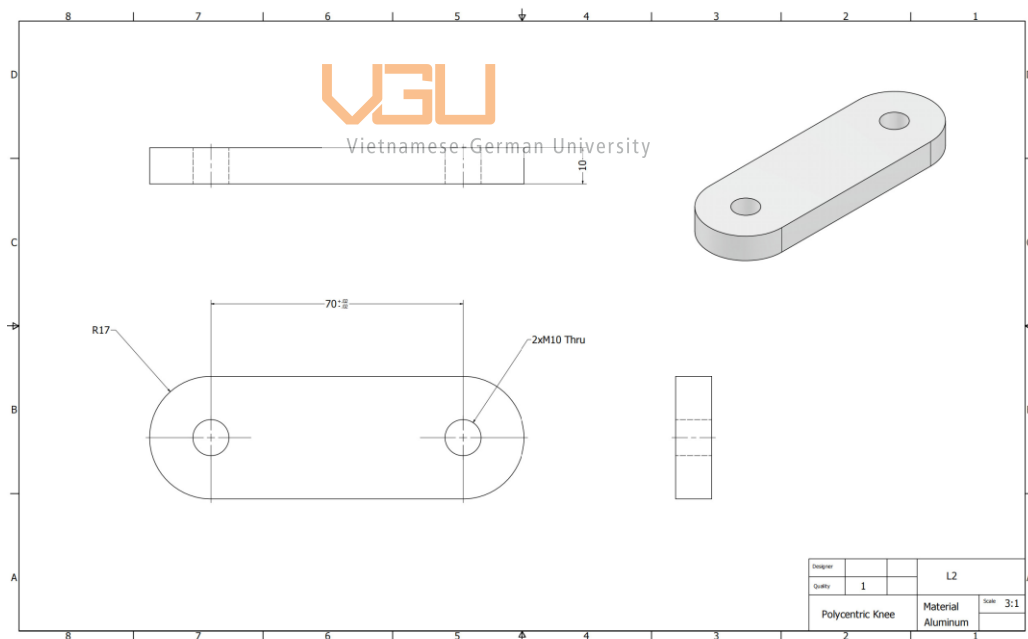


Figure 47 Part L2 Drawing

Part L3

Part L3 shown in **Figure 48** serves as the base for the 4 bar integrated with a slot that allows the leg to rotate 90 degrees, creating comfort for the patient when sitting on a chair. The design of this part is one of the deferential designs (parametric explain in chapter 4).

Following the requirement, the centroid will be a 9mm posterior to the load line shown in **figure 48** below.

With 4 M5 screw holes on top to connect to the Pyramid Adapter that connect to the residual limb socket. **Figure 48** shows the L1 part in the inventor show and **figure 49** shows the L1 drawing.

Material: Aluminum

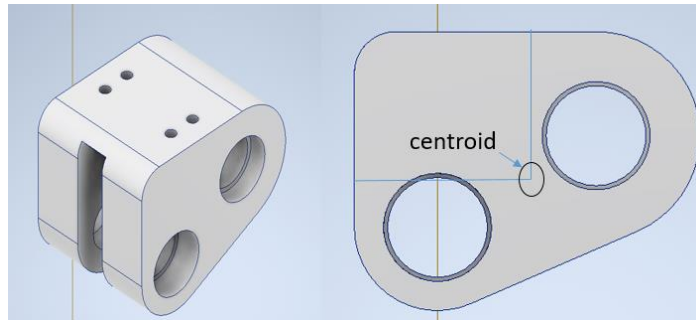


Figure 48 Part L3 in Inventor

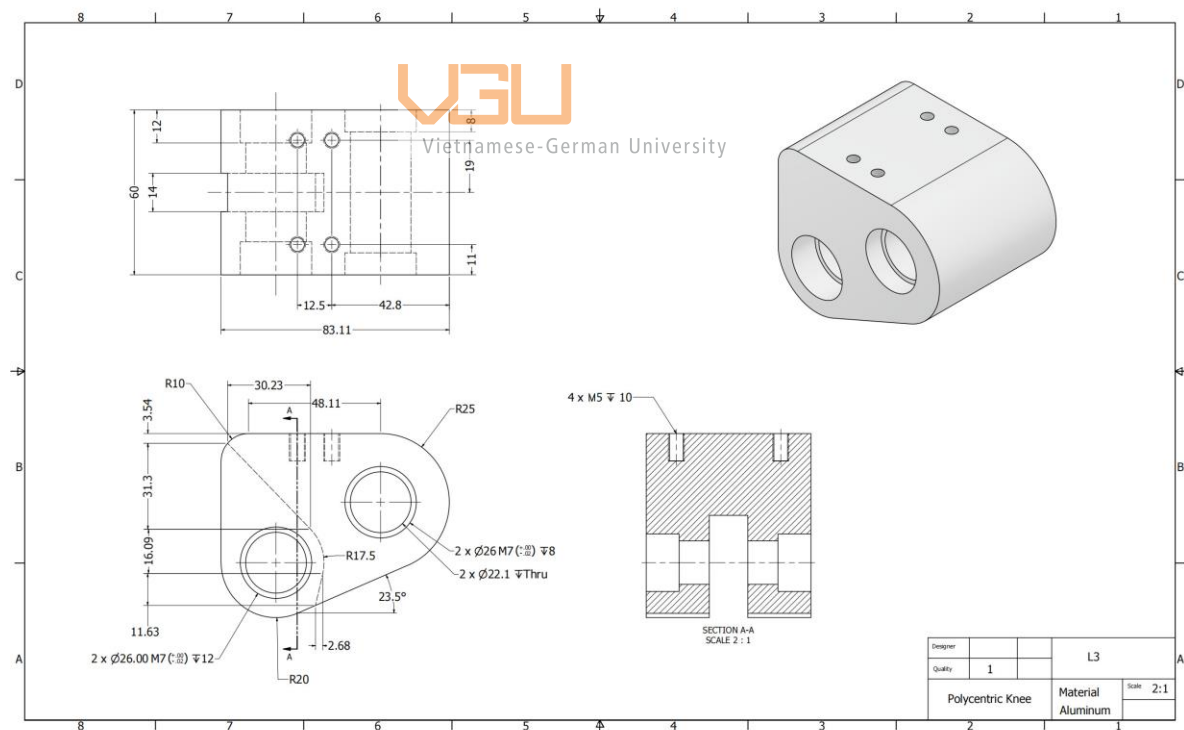


Figure 49 Part L3 Drawing

Part L4

L4 serves as Rocker Link for the 4-bar mechanism shown in **figure 50**. This part connects L1 to L3 with a shaft and bearing. **Figure 50** shows the L4 part in the inventor show and **figure 51** shows L4 drawing.

Material: Aluminum

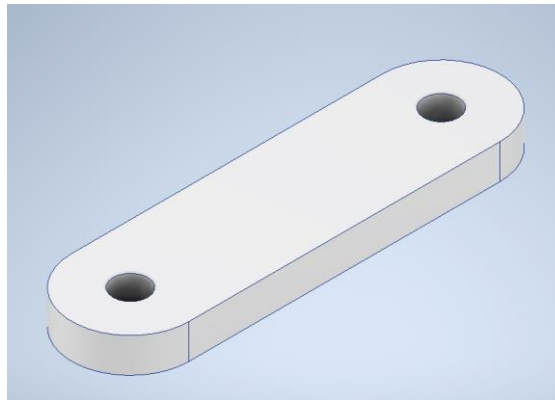


Figure 50 Part L4 in Inventor

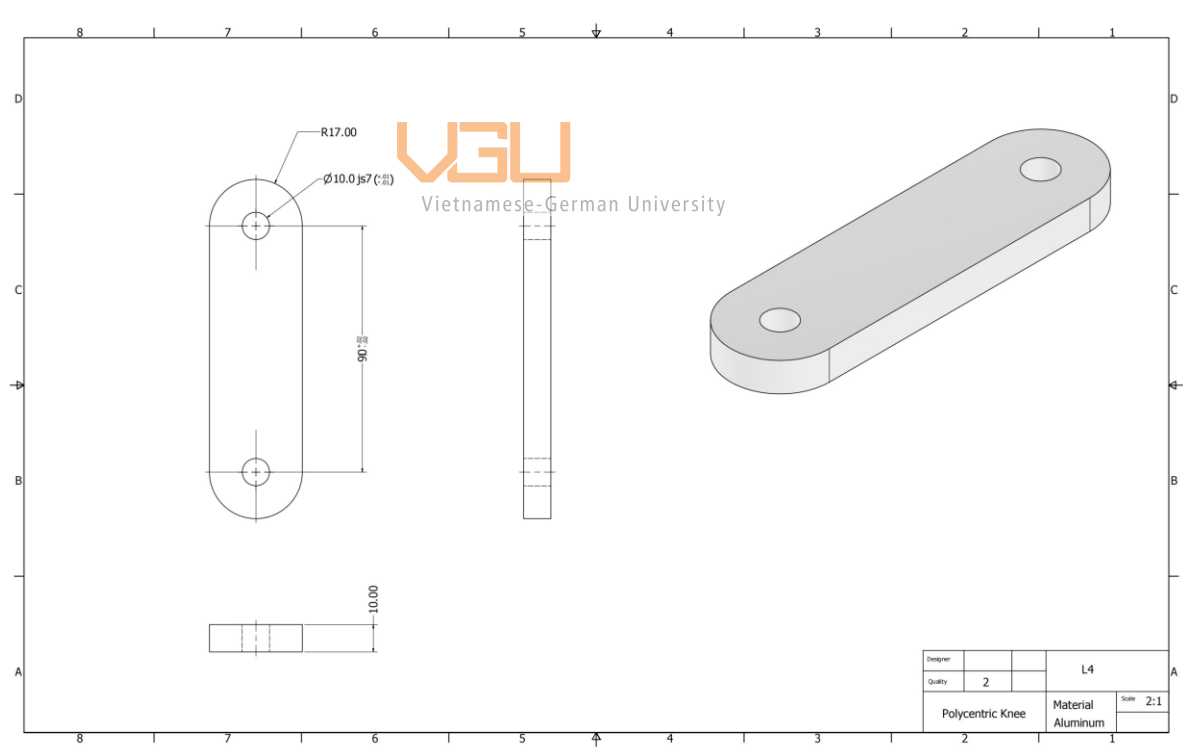


Figure 51 Part L4 Drawing

Inner shaft

Connect parts L1, L3, and L4 together. **Figure 52** shows the Inner shaft part in the inventor show and **figure 53** shows the Inner shaft drawing.

Material: stainless steel 304

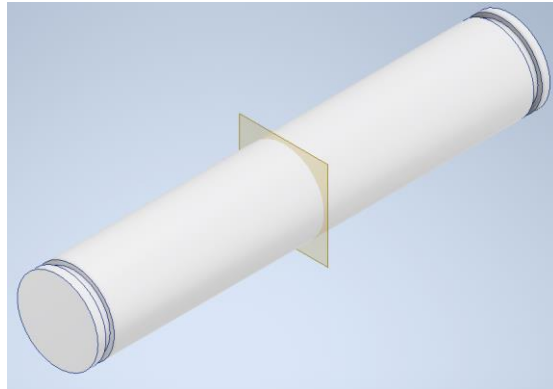


Figure 52 Part inner shaft in Inventor

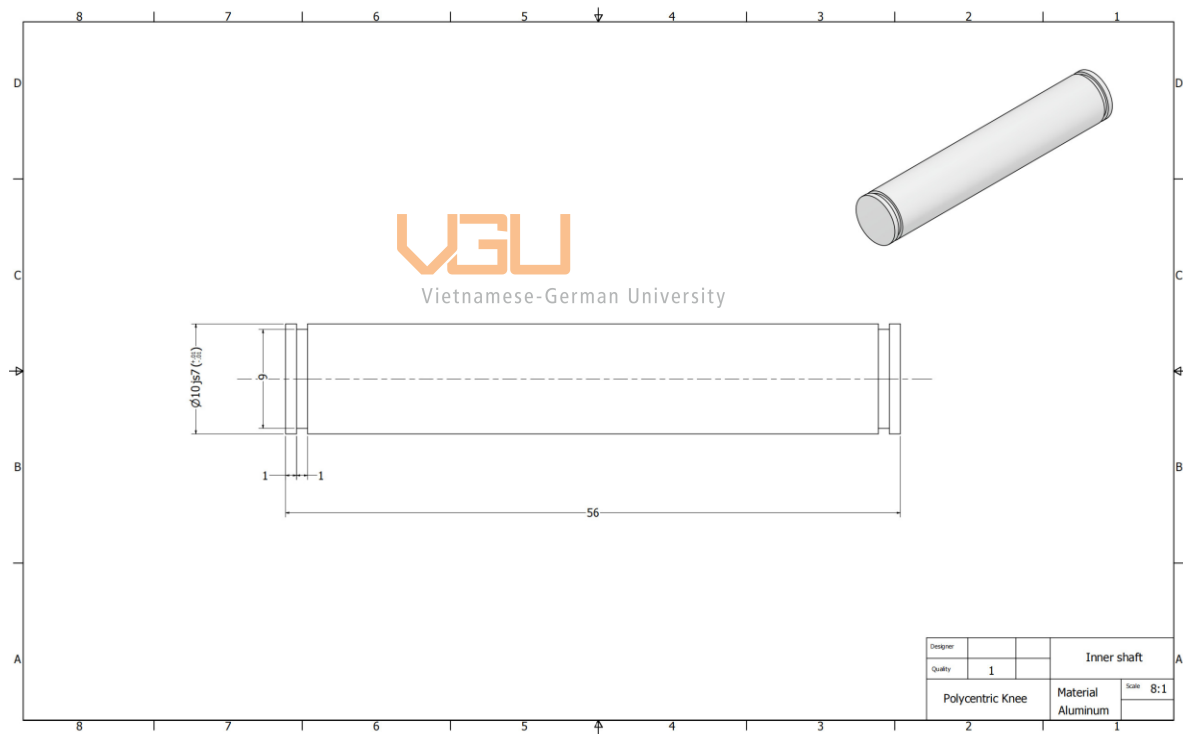


Figure 53 Part inner shaft Drawing

Outer shaft

Connect parts L1, L3, and L2 together. **Figure 54** shows the outer shaft part in the inventor show and **figure 55** shows the outer shaft drawing.

Material: stainless steel 304

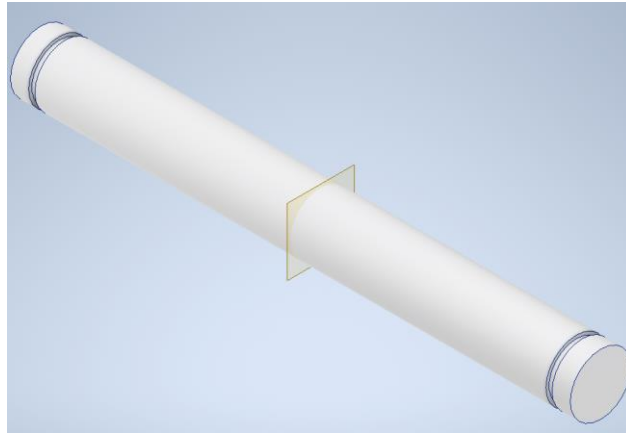


Figure 54 Part outer shaft in Inventor

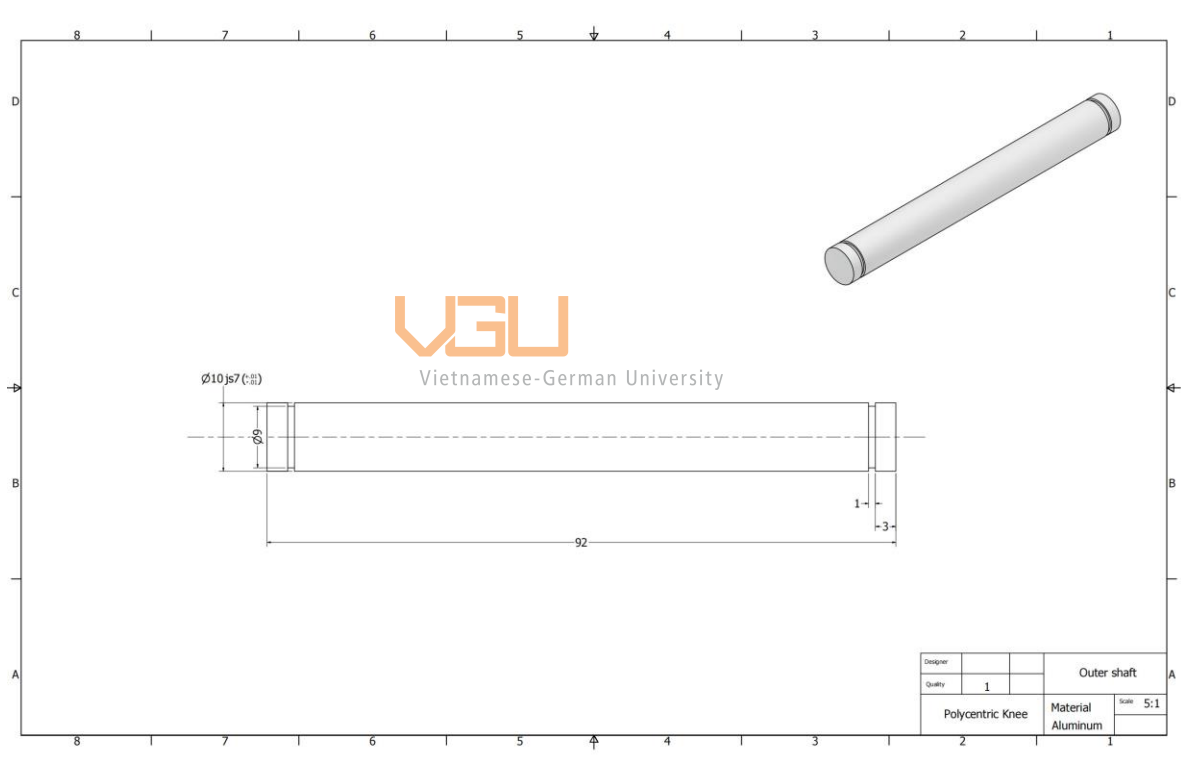


Figure 55 Part outer shaft drawing

Shank Part

The shank is an aluminum pipe that is directly connected to the 4-bar knee using a pipe coupling. **Figure 56** shows the shank part in the inventor show and **figure 57** shows the shank drawing.

Material: Aluminum

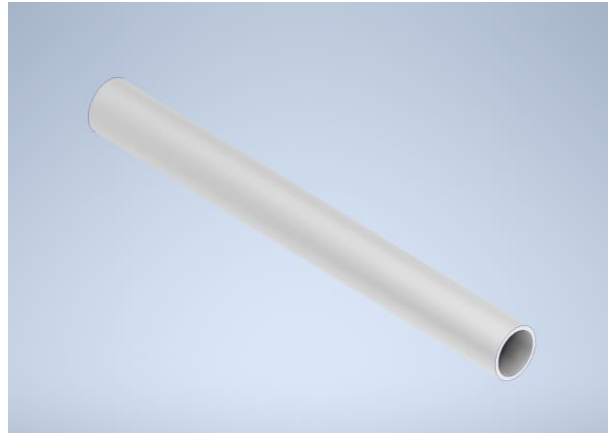


Figure 56 Part shank part in Inventor

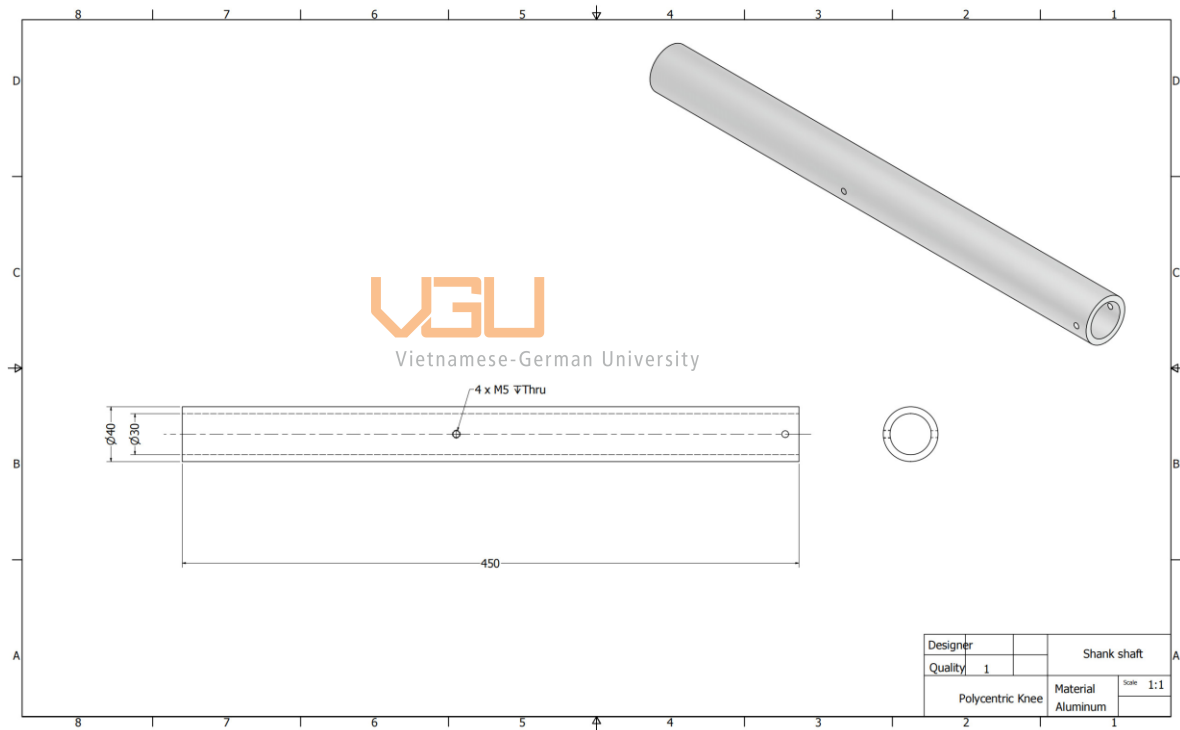


Figure 57 Part shank part drawing

COUPLER

There are two types of coupling widely used in prosthetics for connection and alignment. In the operating process, many problems can occur if the part is carelessly installed, which can result in misalignment and affect user safety.

There are two types of coupling used:

- Pipe Flange: **figure 58** this coupling can easily be tightened with a bolt and nut. One single screwing action is needed to tighten the connection and keep the two parts in place. This makes this easy to assemble. However, it lacks the capability to ground the part into one direction, so we select this to connect the shank to the 4-bar structure.

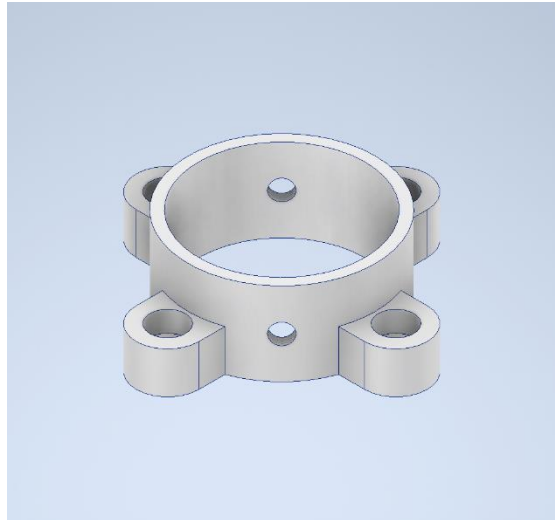


Figure 58 Pipe Flange coupling in Inventor

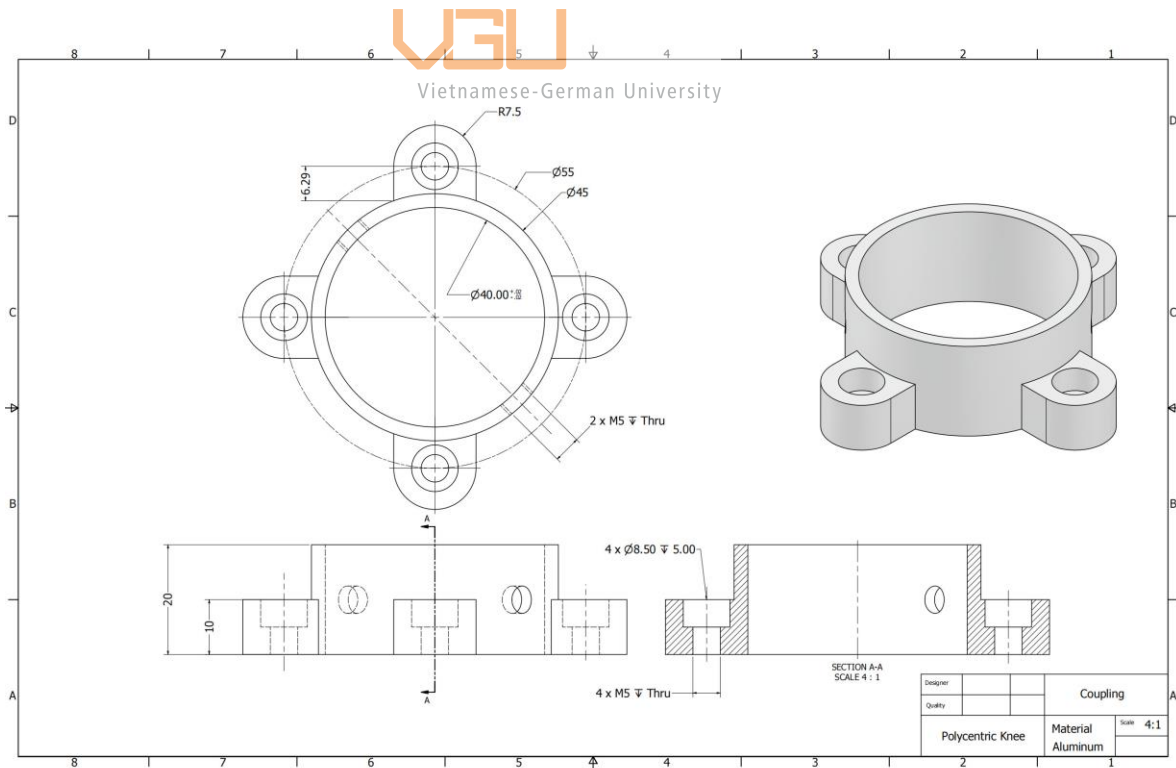


Figure 59 Pipe Flange coupling drawing

- Pyramid Adapter: **figure 60** uses a pyramid of male and female connectors with a four-bolt screw directly to the four sides of the male pyramid adaptor. The said adapter makes it easier to tighten the joint and is more suitable for thigh, knee, and foot orientation. So the pyramid adapter is selected to connect the L3 part with the thigh.

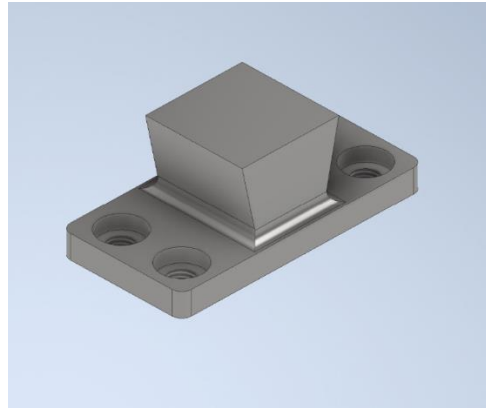


Figure 60 Pyramid adapter in Inventor

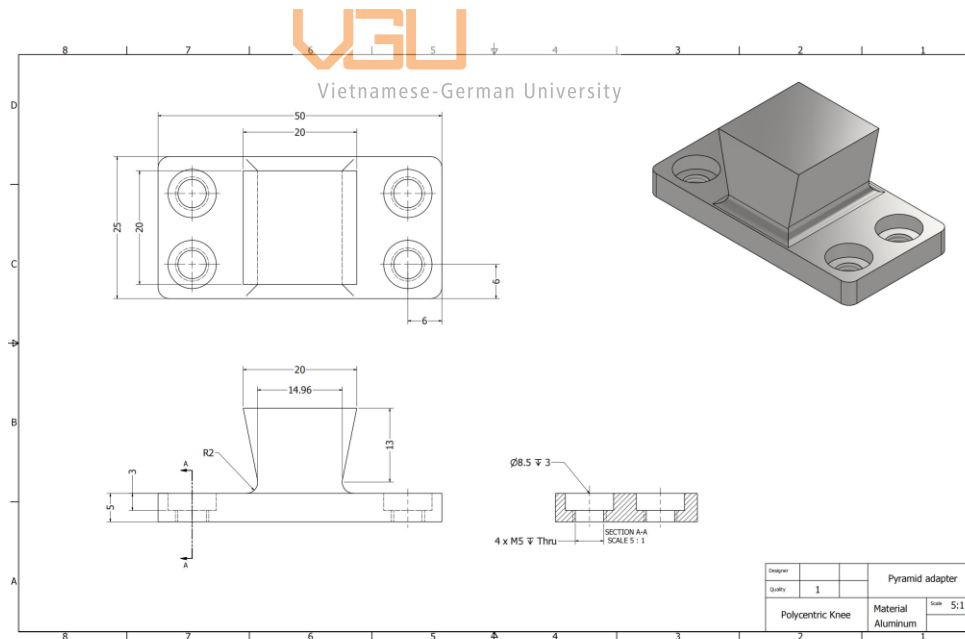


Figure 61 Pyramid adapter drawing

Thrust bearing

To locate and constrain the L3 bar in place

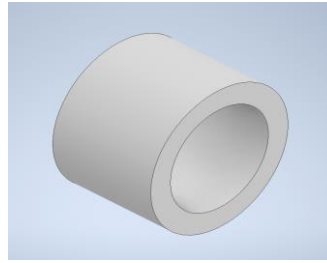


Figure 62 Thrust bearing in Inventor

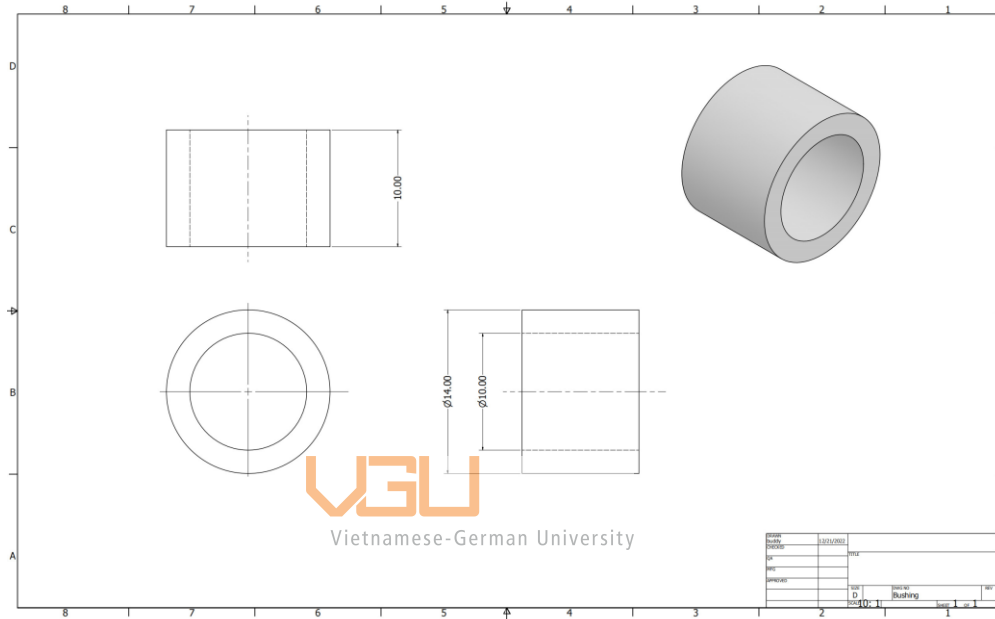


Figure 63 Thrust bearing drawing

Non-Manufacture part

Bolt

We choose bolt for our coupling design with the Proof Loads standard **table 5** below

Proof Loads for Standard Pitch Bolts (Kg's)			
Size	Grade 8.8	Grade 10.9	Grade 12.9
M5	820	1,180	1,380
M6	1,160	1,670	1,950
M8	2,120	3,040	3,550
M10	3,370	4,810	5,630
M12	4,890	7,000	8,180
M16	9,100	13,000	15,200
M20	14,700	20,300	23,800
M24	21,200	29,300	34,200
M30	33,700	46,600	54,400

Table 5 Proof Loads standard table (bolt-grades (fixingsmegastore.co.uk))

Know that with M5 bolt at Grade 8.8 can already handle up to 8200N (820kg) while the maximum load on the prosthetic is estimated at 2800N => Bolt size M5 is chosen for the Coupling Design.

Coupling-L1 Bolt

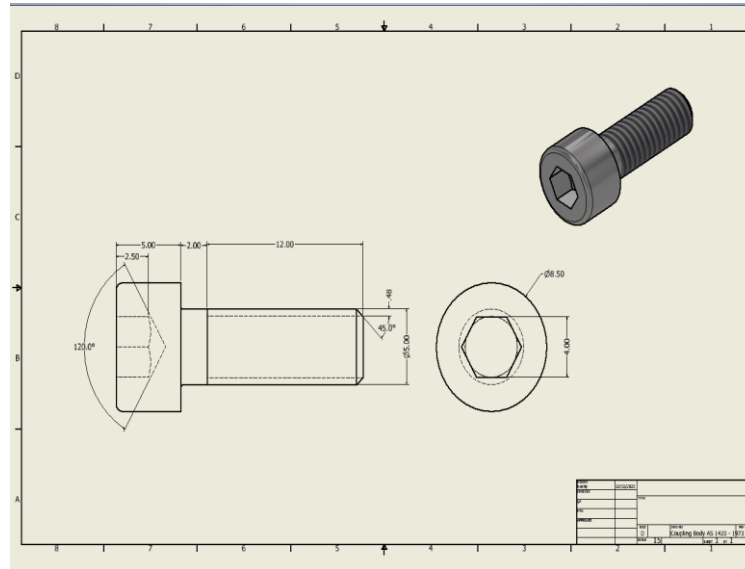


Figure 64 Coupling-L1 drawing

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Coupling-Shank Bolt

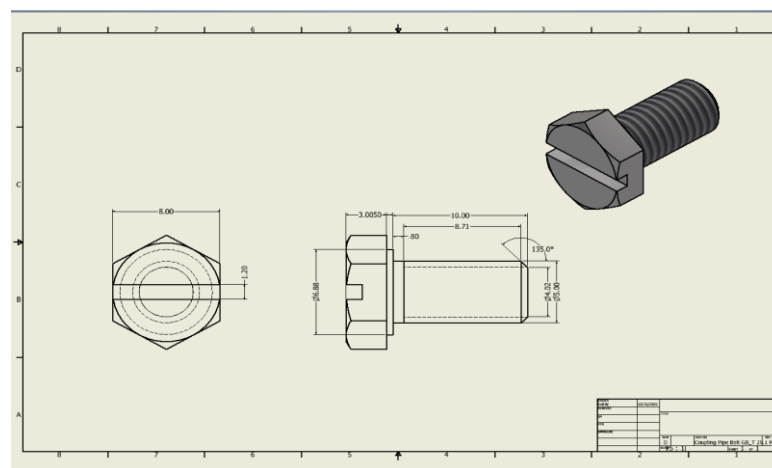


Figure 65 Coupling-Shank Bolt drawing

M8 Retainer

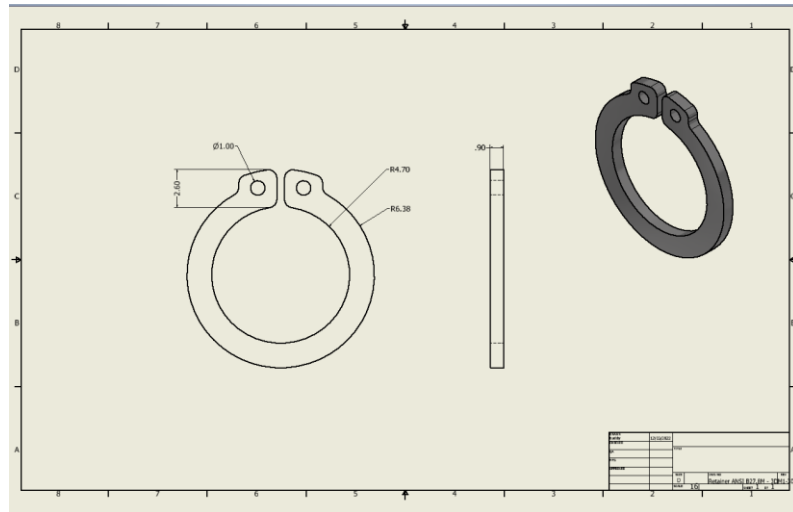


Figure 66 M8 retainer drawing

M22 Washer

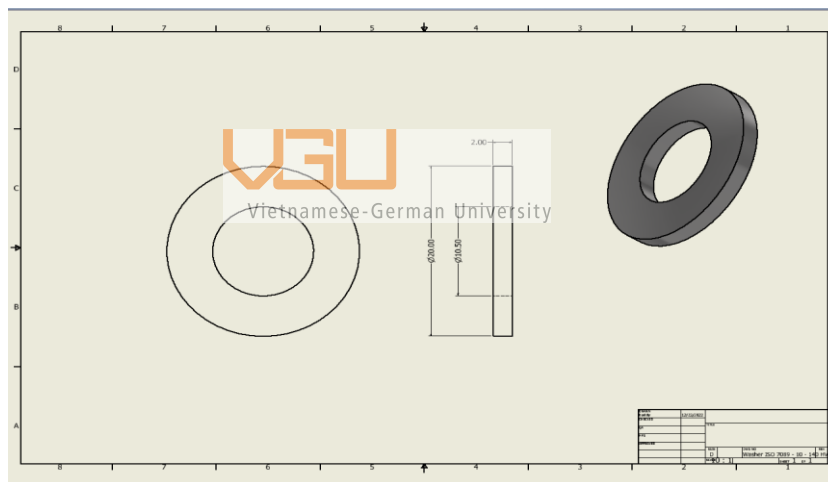


Figure 67 M22 Washing

Pyramid Adapter Bolt

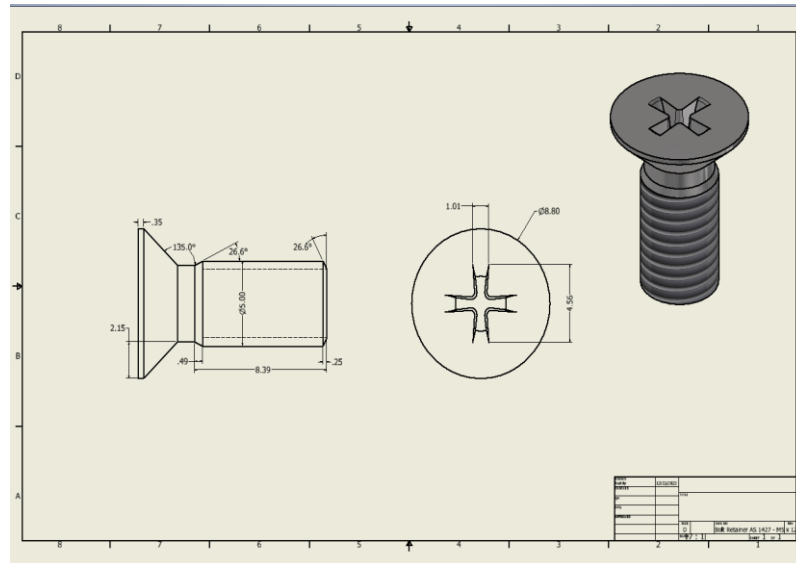


Figure 68 Pyramid Adapter Bolt drawing

Bearing NSK 6000ZZ

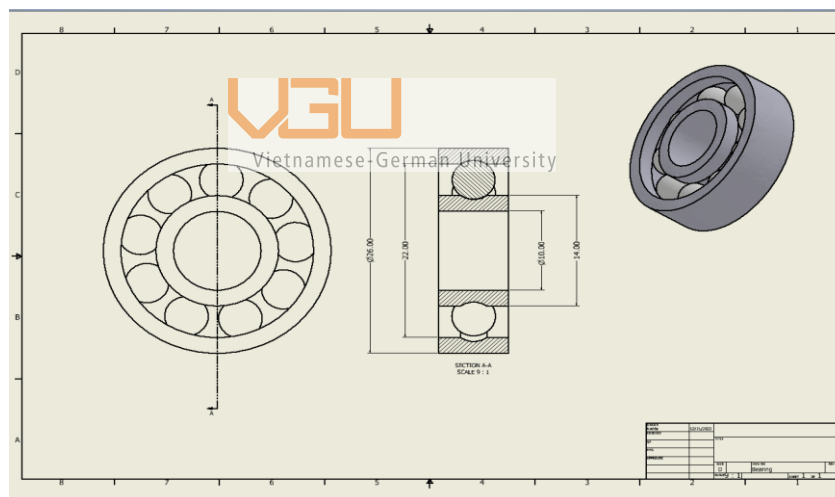


Figure 69 Bearing NSK 6000ZZ drawing

Tolerance and type of fits

First, for parts to be assembled and function correctly, we need to determine the tolerances of the parts as well as how to assemble them. The following list contains details and parts followed up with the reasons for the selection of the required tolerances of each part.

Bearing

To understand bearing tolerances well, first, we need to define the concept of mounting types and how and why to choose the right type for the part. Select mounting by combining different tolerance zones between the hole and shaft with the same accuracy grade or different accuracy grades. In order to select the assemblies, it is necessary to follow the preferred mounting with regard to the bearing type so that the product can be homogenized, the basis tolerance system helps control the upper and lower deviation for the fit and prioritize not changing dimensions unnecessarily because it can reduce machining costs as well as the number of tools and error-checking equipment.

There are two types of tolerance systems for choosing Tolerance for bearing: The shaft system and the Hole system, demonstrated in **figure 70**.

1. Hole Basis System is considered when we want to adjust the fit by changing the dimension of the shaft while keeping the hole dimension fixed. [11]
2. Shaft Basis System is considered when we want to adjust the fit by changing the dimension of the hole while keeping the shaft dimension fixed [11]

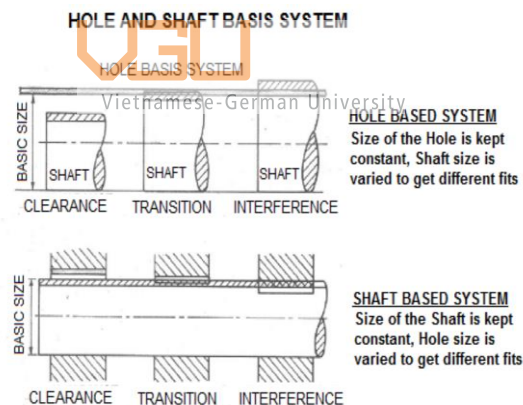


Figure 70 Hole and shaft basic system

List of Tolerance Affects the Bearing design

Bearing Relation to the Body L3: Since the bearing should not be changed, we consider this as a Hole Basis System

Bearing relation to the shaft: Since the bearing should not be changed, we consider this a Shaft Basis System

In this design, we select the bearing 6000zz of NSK to have Specification

主要寸法 (mm)	基本定格荷重 (N)				係数		許容回転数 (min ⁻¹)			呼び番号					
	d	D	B	r	C _r	C _{0r}	C _r	C _{0r}	f ₀	グリース潤滑	油潤滑	開放形	シールド形	シールド形	
10	19	5	0.3		1 720	840	175	86	14.8	34 000	24 000	40 000	6800	ZZ	VV
	22	6	0.3		2 700	1 270	275	129	14.0	32 000	22 000	38 000	6900	ZZ	VV
	26	8	0.3		4 550	1 970	465	201	12.4	30 000	22 000	36 000	6000	ZZ	VV

Figure 71 Bearing specification (Source :Skf Official Site)

With the bearing tolerant Normal $d=10^0_{-0.008}$, $D=26^0_{-0.009}$

The design of then there will be separate into 2 frames.

Frame 1 of reference for the body and the bearing

Figure 72 show the CAD picture of the body and where to assembly bearing



Figure 72 CAD of the hole to assemble bearing

The inner ring of the bearing fits between relative fits and tight fits to the shaft, While the outer ring of the bearing fits relative to the abutment edge.

The shaft basis mounting system is selected (because the bearing is unchanged). Shaft mounting is a mounting system in which clearances and redundancies are created by combining holes with different tolerance zones with a shaft with a single tolerance zone. [11]

The outer ring of the bearing is relative to the body, so it is affected by the load on => cyclic load (The loads acting on the bearing are radial and fixed loads)

D= 22.1mm

Used for parts that are rarely disassembled and are tight attached to the body part Abutment tolerance => N7 is selected

$$D=22.1N7=\text{Ø}22.1_{-0.021}^0$$

Frame 2 of reference 1 relative to the bearing and shaft

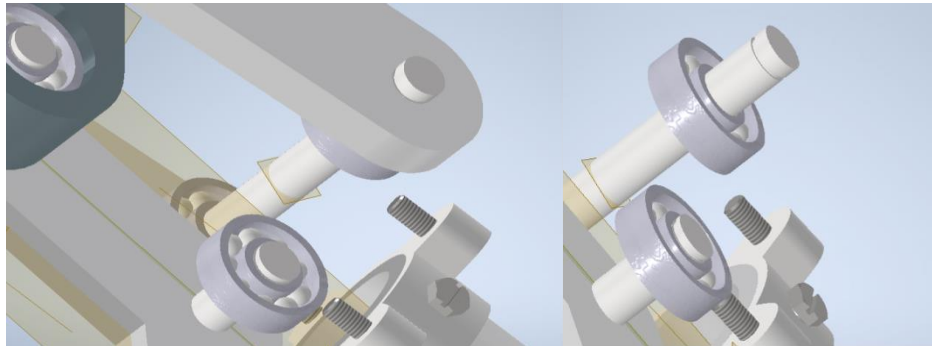


Figure 73 CAD of the shaft and bearing

The inner ring of the bearing with the shaft affected by is only applied force to a part of the bearing. The load on the inner ring => load is local. (The loads acting on the bearing are radial and fixed loads) The hole basis mounting system is selected (because the bearing is unchanged) hole mounting system is a mounting system in which clearances and redundancies are created by combining a shaft with different tolerance zones with a hole having a single zone [11].

With $d= 10\text{mm}$

Commonly used to bear the load of the shaft, so it is necessary to fix the joints tightly nominal size = 10mm to make the shaft fix with the bearing Assembled with transition fit so the Js7 tolerant of the shaft is selected => shaft diameter $d=10Js7=\text{Ø}10_{-0.009}^{+0.009}$

Shaft vs L2/L4



Figure 74 CAD of L2/L4 hole

The shaft connects the L2 and L4 with the body. The transition fit is selected because the shaft is only applied force to a part of the bearing (local)

In this part, a shaft mounting system is selected (because the bearing is unchanged). Shaft mounting is a mounting system in which clearances and redundancies are created by combining holes with different tolerance zones with a shaft with a single tolerance zone.

L2 and L4 hole is selected the Transition fits $D=10J_7=\Phi 10_{-0.009}^{+0.009}$

The remaining details are not affected by direct force and do not need high accuracy, so they will be determined together with IT12 tolerances.

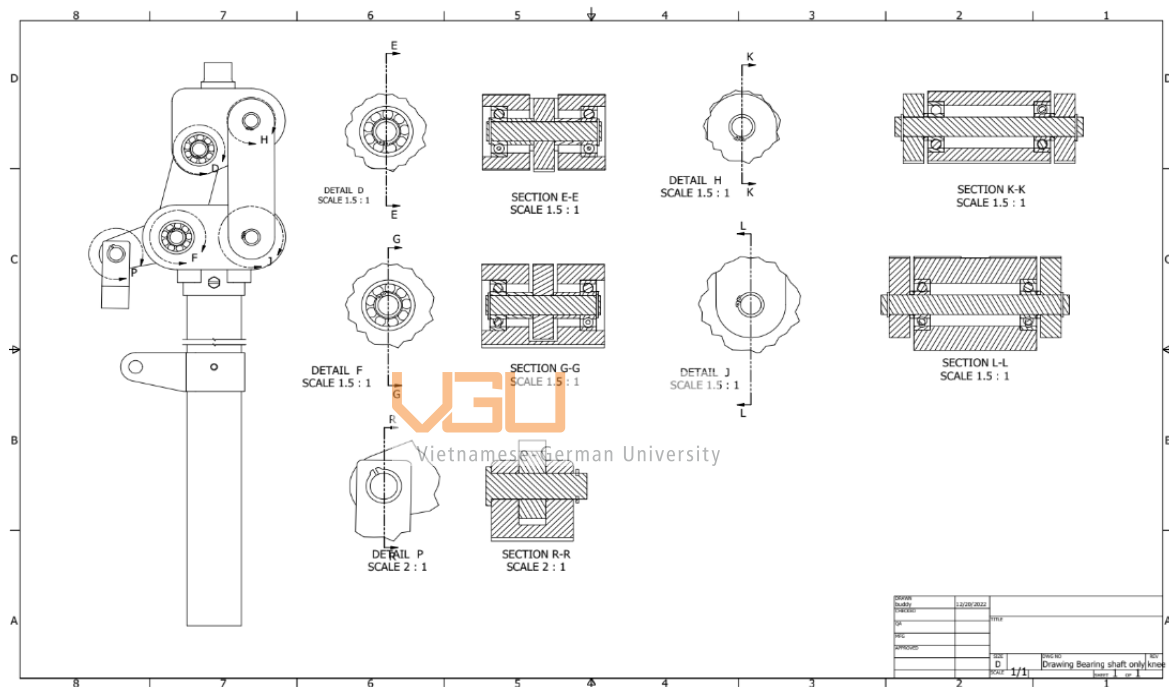


Figure 75 Assembly Drawing

Static mechanical Analysis

Finite Element Analysis

Finite Element Analysis: Modeling a component's or assembly's behavior under predetermined conditions in order to analyze it using the finite element method (FEA) is known as finite element analysis (FEA). Engineers use FEA to simulate dynamic behavior, without the need for actual prototypes. [12]

FEA is using numerical method for forecasting how an assembly or component will function under specific conditions. It serves as the foundation for current simulation software, which

assists engineers in identifying weak points and areas of strain in their designs. The results of a FEA-based simulation are frequently displayed using a color scale that, for instance, shows the pressure distribution over the item. [12]

Von Mises Stress

A test to determine if a material may give or fracture. It is mostly used to ductile materials. A material will yield if its von Mises stress under load is equal to or greater than its yield limit under simple tension, according to the von Mises yield criterion.

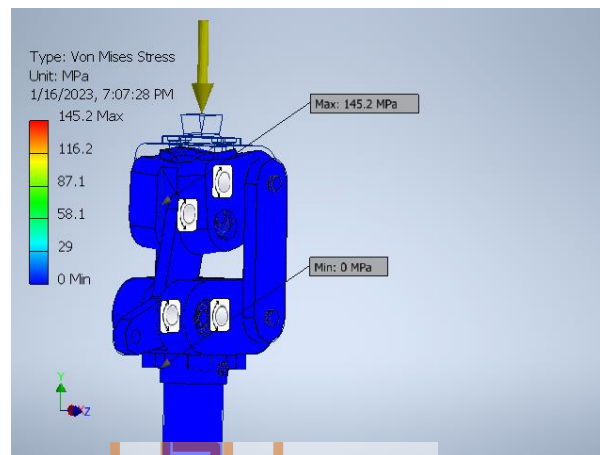


Figure 76 Von Mises stress analysis
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The result of von Mises stress is show in **figure 76**, on this figure the maximum stress is 145.2Mpa when we applied force on the adapter we can see there a small mesh region of the region shifted and the adapter a deform. The outcome of this Von Mises Stress Analysis may be used to inform the design engineer of the present design's weak points. [12]

Displacement

Computing the displacement field within a material exposed to external forces is the primary goal of the displacement based finite element method. As the load is applied, every point in the solid shifts. [12]

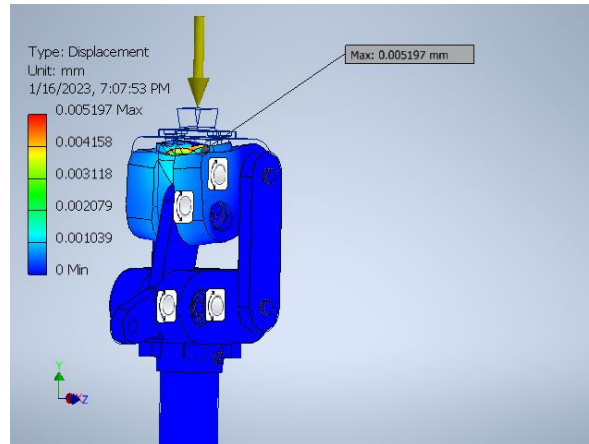


Figure 77 Displacement analysis

A load of 2700N is placed on the roller in this situation. We can see the pyramid adapter and the L3 part are shifted downward to 0.005197mm.

1st principal stress and The 3rd principal stress

After Displacement simulate we see the 1st principal stress and The 3rd principal stress of design.

"The 1st principal stress provides the value of stress that is perpendicular to the plane in which the shear stress is 0. The 1st principal stress provides you the highest tensile stress created in the component due in part to the loading situations."

"The 3rd principal stress acts normal to the plane in which shear stress 0. It provides the maximum compressive stress created in the component during loading."

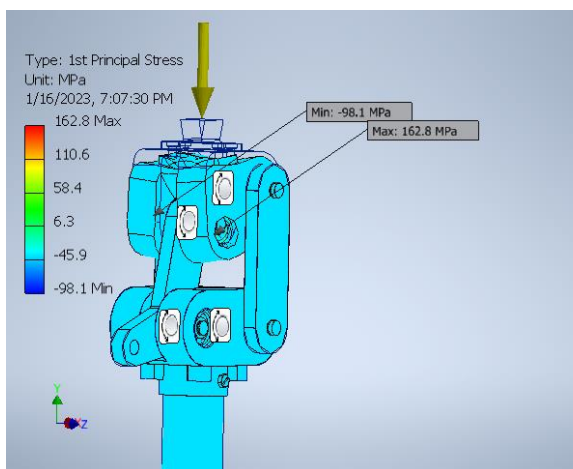


Figure 78: 1st principal stress analysis

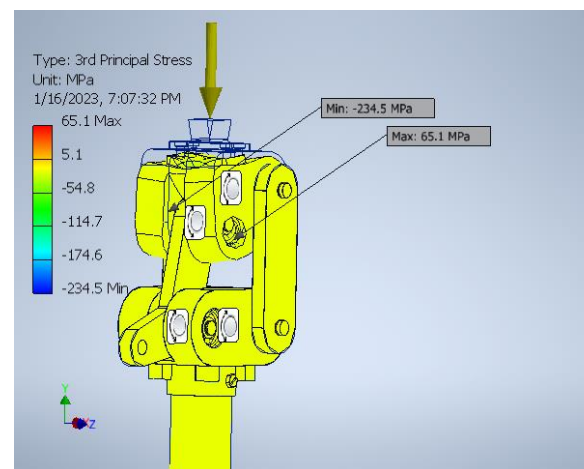


Figure 79: 3rd principal stress analysis

As the result has been shown The part which affected most under the estimated 2800N are the deviation on part L3 and shaft. Hence in the worst-case scenario, the Prosthetic leg only needs minor maintenance and will not be disassembled after the collision.

Chapter 6: Manufacturing

Introduction

In this chapter, we will show the part of the 4-bar prosthetic when the prototype is made and how the assembly of the structure together, bill of material, and mass is also shown.

Manufacture part

Part L1



VGU
Figure 80 Part L1
Vietnamese-German University

Function:

- Main part of the structure
- Connect with the shank take on the transfer movement form the hip to the shank though L3

Material: 6061 Aluminum

Volume: 204064.017 mm³

Mass: 0.55kg

Part L2



Figure 81 Part L2

Function:

- Main part of the structure
- Connect with L3 with L1 take on the movement transfer from L3 to L1

Material: 6061 Aluminum

Volume: 38108.406 mm³

Mass: 0.12kg

Part L3

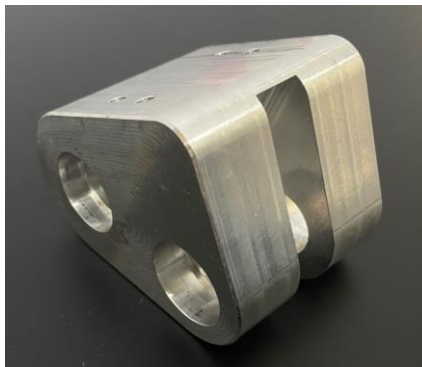


Figure 82 Part L3
VGU
Vietnamese-German University

Function:

- Main part of the structure
- Connect with the upper body through add perter

Material: 6061 Aluminum

Volume: 201901.362 mm³

Mass: 0.55 kg

Part L4



Figure 83 Part L4

Function:

- Main part of the structure
- Connect with the L3 and L1 transfer from L3 to L1 also limited the movement at the definite angle

Material: 6061 Aluminum

Volume: 40580.630 mm³

Mass: 0.11kg

Inner shaft



Figure 84 Inner shaft

Function:

- Main part of the structure
- Connect part together

Material: 304 stainless steel

Volume: 4368.385 mm³

Mass: 0.035 kg

Outer shaft



Figure 85 Outer shaft

Function:

- Main part of the structure

- Connect part together

Material: 304 stainless steel

Volume: 7195.818 mm³

Mass: 0.058kg

Pyramid adapter

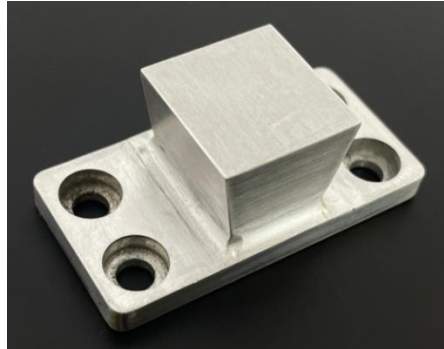


Figure 86 Pyramid adapter

Function:

- Main part of the structure
- Connect to the upper body thigh and socket

Material: 6061 Aluminum

Volume: 10577.138 mm³

Mass: 0.029kg

Shank part



Figure 87 Shank part

Function:

- Main part of the structure
- Connect with the 4 bar structure replicate the leg tibia bone

Material: 6061 Aluminum

Volume: 247005.856 mm³

Mass: 0.67kg

Pipe coupling



Figure 88 Pipe coupling

Function:

- Sub part of the structure
- Connect the shank with 4 bar structure

Material: 6061 Aluminum

Volume: 11838.103 mm³

Mass: 0.032kg

Thrust bearing



Figure 89 Thrust bearing

Function:

- Sub part of the structure
- Mount between shaft and body to locate and constrain the L3 bar in place

Material: 6061 Aluminum

Volume: 753.982 mm³

Mass: 0.002kg

6000z bearing



Figure 90 6000z bearing

Function:

- Main part of the structure
- Attach with a body and connect with a shaft to connect part of the structure together

Material: 6061 Aluminum

Volume: 201901.362 mm³

Mass: 0.019 kg

List of material

No	Parts	Material	Quantity	Weight (kilograms)
1	L1	Aluminum	1	0.55kg
2	L2	Aluminum	2	0.12kg
3	L3	Aluminum	1	0.55 kg
4	L4	Aluminum	1	0.11kg
5	Inner Shaft	304 stainless steel	2	0.035 kg
6	Outer Shaft	304 stainless steel	2	0.058kg
7	Pyramid Adapter	Aluminum	1	0.029kg
8	Shank Pipe	Aluminum	1	0.67kg
9	Pipe Coupling	Aluminum	1	0.032kg
10	Thurs bearing	Aluminum	4	0.002
11	Bearing NSK 6000ZZ	Steel	8	0.019 kg
12	Washer M22	Steel	8	N/A
13	Retainer/Snap ring/Circle M8	Steel	8	N/A
14	Adapter Bolt	Steel	4	N/A
15	Coupling Bolt	Steel	4	N/A
16	Shank-Coupling Bolt	Steel	2	N/A
Total				2.57kg
Total cost (VND)				5.000.000vnd

Table 6 List of material

Assembly

Figure 91 is an expose assembly drawing of the prototype and explain step by step below by inventor.

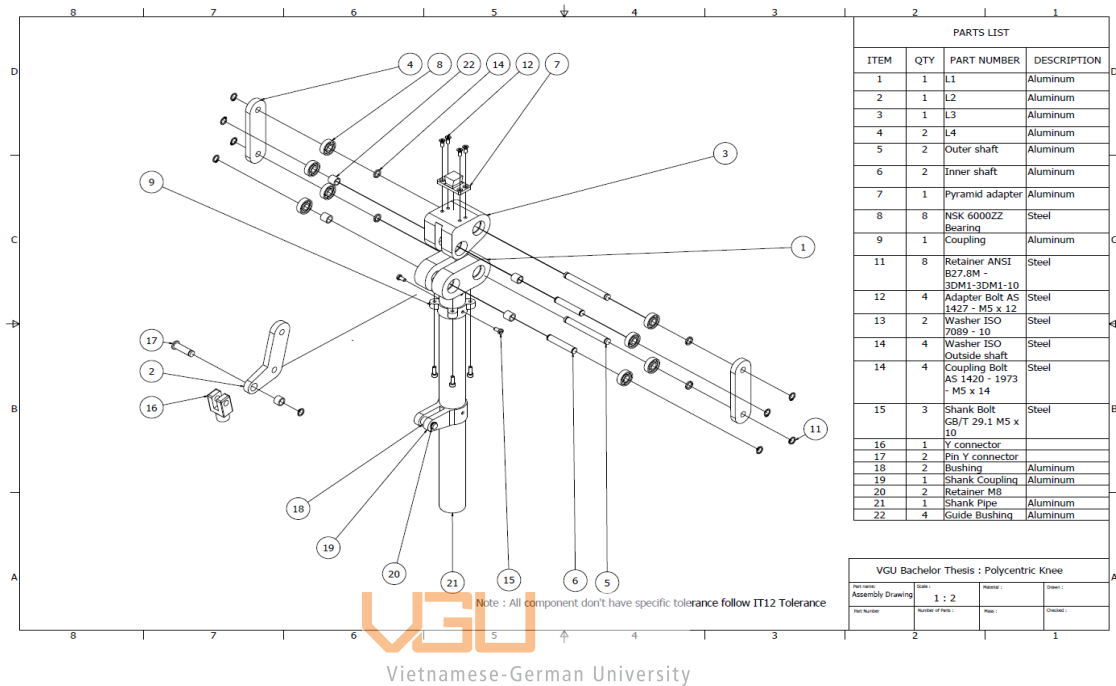


Figure 91 Expose assembly drawing

Firstly, we assembly the shaft with the thrust bearing show in **figure 92**.

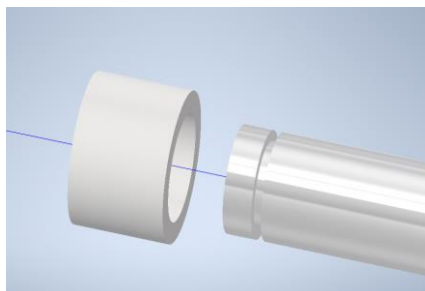


Figure 92 Shaft with the thrust bearing assembly

Then we assembled the bearing to the body because the tolerance of the bearing is a tight fit, so to do this, we used a rubber hammer and heat extension (heating the hole for mounting the bearing) to the assembly. Loctite 680 glue is added to the mounting hole, and the L2 connects

L3, L1 by a shaft with M22 Washer **figure 93**. The shaft is then locked by an M8 retainer **figure 94**.

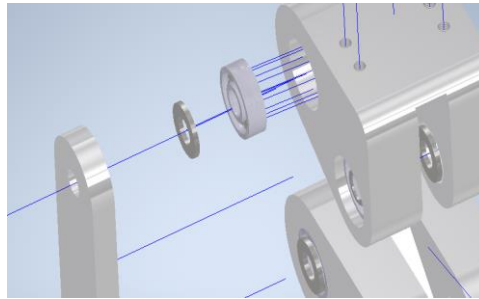


Figure 93 Expose of L2 and L3 connect by L1



Figure 94 Shaft locked by an M8 retainer
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Finally, the pyramid adapter connects L3 with four hexagon bolts, and the coupling connects the shank with L1 by four M5 bolts **figure 95**.

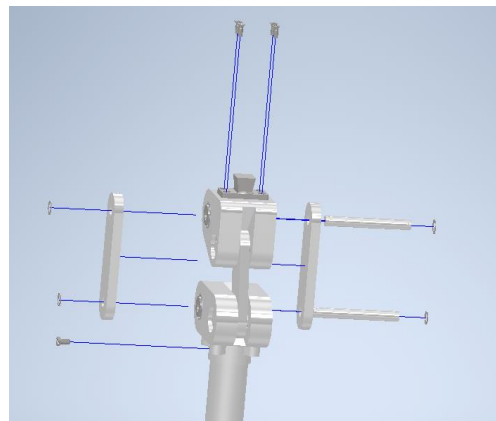


Figure 95 Assembly of adapter and coupling

Figure 96 is a real prototype part and after assembly prototype.



Figure 96 All part and real prototype after assembly

Chapter 7: Testing

The polycentric leg is going to be tested during the swing phase with three categories: Gait cycle, Toe clearance, Acted Force and speed. All the testing will be used with the same equipment and using WINalyze (motion tracking and analysis) software testing format will be done as follow:

A system consists of an infrared camera that detects the marker that is placed on the primary joint of the knee to collect the data for position and direction.

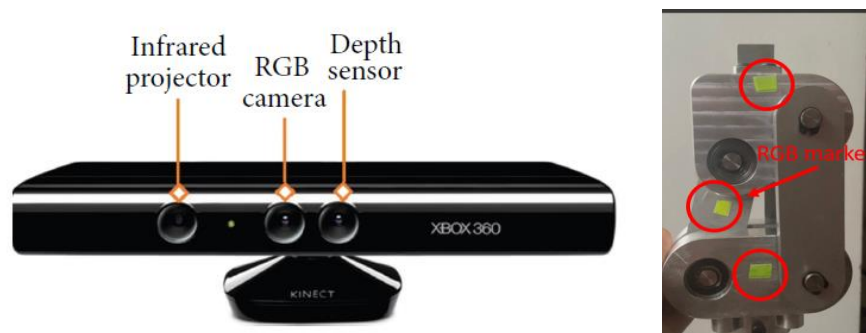


Figure 97 Equipment and RGB maker point

RGB marker will be attached to the joint of the polycentric knee and a system of infrared camera will detect and analyze the movement of full gait cycle of movement.

Gait Cycle and Movement Testing

First, we begin with the gait cycle. We got the gait cycle angle of the Swing phase simulated in table 4 in chapter 4. This number already projects with the real human angle, so we just need to compare this simulated angle with the prototype angle to find out if the prototype is fully functional or not.

Gait Cycle angle in swing phase									
Frame	1	2	3	4	5	6	7	8	9
Angle	82.3	75.2	70.7	65.2	60.4	55.2	50.2	45.9	40

Table 7 Gait Cycle angle in swing phase

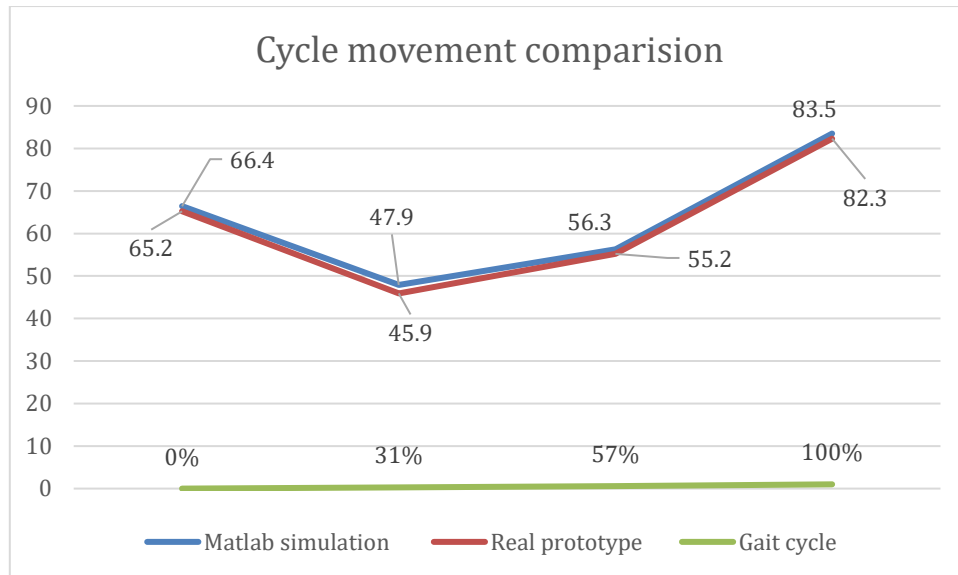


Figure 98 Gait cycle comparison simulation vs real prototype

Table 7 shows us the angle of L1(ground) and L2 (crank) θ_2 . This number is compared with **table 4** simulation and demonstrated in a chart in **figure 98**. As you can see, the number is almost the same as the simulated number and has a negligible error, so the prototype is fully functional.



Toe clearance testing

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To calculate minimum toe clearance following. We have to calculate the distance between the toe point of the leg and the ground [9]. In order to do that, we use the same method for the gait cycle test with the infrared camera.

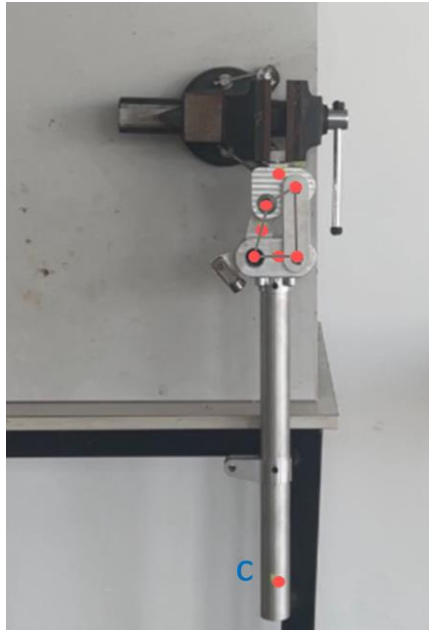


Figure 99 Prosthetic knee at stance



Figure 100 Prosthetic knee at the end of swing

In this test, the full cycle of the swing phase is conducted, the toe point of the leg can be considered point C' lowest RBG point of the leg **figure 100** and the ground is point C **figure 99** with the motion tracking software we have the result of the distance of C and C' is 12.8mm.



Vietnamese-German University

Four-bar knee configurations	Proteval	Ottobock 3R36	Ottobock 3R55	Ottobock 3R70	Prototype	IITM knee
Toe clearance measure in mm	2.2	2.4	19.4	2.8	12.8	11

Table 8 Comparison of toe clearance 1

The purpose of our design is to increase toe clearance. As you can see from **table 8**, this digit is increased compared to another market 4 bar knee. So the design is improving the toe clearance.

Speed and force at heel contact

At heel contact, the knee angle is at 90 degrees with the ground. With this test, the infrared camera and software are just used to provide the input variables properties the resulting output is calculated manually.

φ_1	φ_2	t_2	t_1	d	m
67°	0°	0s	0.84s	0.6m	2.57kg

Table 9 Input variables properties 1

Speed

to calculate the speed of the swing we using the energy conservation formula

$$W_{kinetic} = \frac{1}{2}mv^2 \quad (7.1)$$

$$W_{kinetic} = mgh \quad (7.2)$$

$$\frac{1}{2}mv^2 = mgh \Rightarrow v = \sqrt{2gh} = 2.07 \text{ m/s} \quad (7.3)$$

Force

To calculate the force, the moment at the 4 knee take into account

$$\vec{M} = \vec{F} * \vec{d} \quad (7.4)$$

To find F we need to find the moment

$$M = I * B \quad (7.5)$$

We got I is a moment of inertia equation (7.6) and B is angular acceleration equation (7.7)

$$I = m * d \quad (7.6)$$

$$B = \omega' = \Delta \frac{d\omega}{dt} \quad (7.7)$$

With angular velocity equation (7.8)

$$\omega = \frac{d\varphi}{dt} = \frac{\varphi_2 - \varphi_1}{t_2 - t_1} \quad (7.8)$$

By step by step with input value that provided by testing with infrared camera the result of F is

$$F = 4N$$

So 4N is the force needed to brake the swing, this result of speed and force can be used for future work and optimization.

To summarize, the prototype is designed and works properly according to requirements, but there are still points that can be improved to make the prototype operate with higher efficiency, especially with force and speed.

Chapter 8: Conclusion

Conclusion

To summarize, the conceptualizing, designing, manufacturing, assembling, and testing stages of the thesis project have been conducted. The prototype is made, and the result meets the requirement given. However, there is still room for further improvement. After the research and implementation of the thesis: "DESIGN, FABRICATION, AND TESTING OF PROSTHETIC LEG". We have achieved the following results:

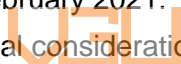
- Understand the situation of the medical equipment manufacturing industry, the opportunities, challenges, as well as inadequacies, and limitations, especially in terms of competition and dependence on raw materials.
- Identify the main functions and technical requirements of a prosthetic leg.
- Provide design options for the Polycentric Knees (4 bar) and the selection of the plan that meets the basic requirements set out earlier for the design of the mechanical part.
- The development direction of the thesis is the premise for the development of prosthetics which in Vietnam are limited and usually depend on foreign.
- Through this thesis, we have obtained theoretical and practical experience in health equipment as well as the prosthetic field.

Future improvement

Further development directions for this thesis can include:

- Designed to make the device more compact and functional such as selecting a more advanced alternative material that can reduce the parameter
- Design the braking system adding to reduce the speed during when swing phase, MR damper is one of the possible options because it can adjust damping force by each phase.
- Optimize the control problem, and improve automation. The design can modify to a computerized 4-bar mechanic with PIC and PID control for reducing the swing speed by a braking system.
- Improve the appearance and style of the prosthetic leg for a better market appeal.

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APPENDIX

q	p	Yc
15	37.0002	21.73941
15.02	37.0202	21.73734
15.04	37.0402	21.73528
15.06	37.0602	21.73322
15.08	37.0802	21.73116
15.1	37.1002	21.72911
15.12	37.1202	21.72706
15.14	37.1402	21.72501
15.16	37.1602	21.72297
15.18	37.1802	21.72093
15.2	37.2002	21.71889
15.22	37.2202	21.71686
15.24	37.2402	21.71483
15.26	37.2602	21.7128
15.28	37.2802	21.71078
15.3	37.3002	21.70876
15.32	37.3202	21.70674
15.34	37.3402	21.70472
15.36	37.3602	21.70271
15.38	37.3802	21.7007
15.4	37.4002	21.69869
15.42	37.4202	21.69669
15.44	37.4402	21.69469
15.46	37.4602	21.69269
15.48	37.4802	21.6907
15.5	37.5002	21.68871
15.52	37.5202	21.68672
15.54	37.5402	21.68474
15.56	37.5602	21.68275
15.58	37.5802	21.68077

Table: Result of p and Yc according to q



Figure: Gait Cycle angle represent in RGB dot in swing phase