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Asymmetric Unilateral Transfemoral Prosthetic Simulator

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by

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ABSTRACT

Transfemoral amputees develop a physical asymmetry because of their amputation, which includes reduced force generation at the knee and ankle, reduced control of the leg, and different mass properties relative to their intact leg. The physical change in the prosthetic leg leads to gait asymmetries that include spatial, temporal, or force differences. This altered gait can lead to an increase in energy consumption and pain due to compensating forces and torques. The asymmetric prosthesis demonstrated in this research aims to find a balance between the different types of asymmetries to provide a gait that is more symmetric and to make it overall easier for an amputee to walk.

Previous research has shown that a passive dynamic walker (PDW) with an altered knee location can exhibit a symmetric step length. An asymmetric prosthetic simulator was developed to emulate this PDW with an altered knee location. The prosthetic simulator designed for this research had adjustable knee settings simulating different knee locations. The prosthetic simulator was tested on able-bodied participants with no gait impairments. The kinetic and kinematic data was obtained using a VICON motion capture system and force plates.

This research analyzed the kinematic and kinetic data with different knee locations (high, medium, and low) and normal walking. This data was analyzed to find the asymmetries in step length, step time, and ground reaction forces between the different knee settings and normal walking.

The study showed that there is symmetry in step lengths for all the cases in overground walking. The knee at the lowest setting was the closest in emulating a normal symmetric step length. The swing times for overground walking showed that the healthy leg swings at almost the

same rate in every trial and the leg with the prosthetic simulator can either be symmetric, like the healthy leg or has a higher swing time. Step lengths on the treadmill also showed a similar pattern, and step length of the low knee setting were the closest to the step length of normal walking. The swing times for treadmills did not show a significant trend. Kinetic data from the treadmill study showed that there was force symmetry between the low setting and normal walking cases. In conclusion these results show that a low knee setting in an asymmetric prosthesis may bring about spatial and temporal symmetry in amputee gait.

This research is important to demonstrate that asymmetries in amputee gait can be mitigated using a prosthesis with a knee location dissimilar to that of the intact leg. Tradeoffs have to be made to achieve symmetric step length, swing times, or reaction forces. A comprehensive study with more subjects has to be conducted in-order to have a larger sample size to obtain statistically significant data. There is also opportunity to expand this research to observe a wider range of kinetic and kinematic data of the asymmetric prosthesis.

CHAPTER 1: INTRODUCTION

Human walking is a complex process involving various muscles and precise neurological synchronization. The legs swing back to front to generate forward motion, and each leg swings 180 degrees out of phase with the other. This gait cycle is altered by amputation, stroke, physical changes, and other forms of physical and neural disorders. The goal of this research is to correct gait irregularities found in transfemoral prosthesis users.

This thesis shows the proof of concept of the theoretical transfemoral prosthesis design based on passive dynamic walkers [39] [87] [88]. Transfemoral amputation and knee dis-articulation [4] are procedures where the person loses the function of the knee and the ankle joints. Transfemoral amputees with a symmetric prosthesis compensate for altered forces with their intact leg, alter their gait in order to walk comfortably, expend higher energy in walking than able bodied and transtibial amputees, and often experience pain in the intact leg and at the hip joints. Thus, there is a need to further refine prosthetic devices.

The design of this prosthesis looks to improve the quality of gait in transfemoral prosthesis users. The model is based on passive dynamic walkers [88], and shows that the shift in knee location decreases the overall prosthesis weight. The reduction in weight combined with a shorter shank swing reduces the energy cost required to walk [41]. To obtain data without relying on a subject with transfemoral amputation, I decided to test it out on able bodied subjects. This resulted in the asymmetric unilateral transfemoral prosthetic simulator.

The prosthetic simulator helps in the study of gait rehabilitation in many ways. The simulator does not require a customized design and can be worn by any able bodied person. This simulator differs a lot from its predecessors because the knee location is below the anatomical position allowing for a simpler design without an offset to accommodate the knee. The shifting of the knee also makes the design lightweight. The problem statement for the design also required a completely passive system which resulted in a low cost design.

The scope of this thesis is to demonstrate the efficacy of the prosthesis with the shifted knee location in real world conditions. The prosthetic simulator is designed to have variable knee locations. The walking behavior of the wearer is compared at every knee location. According to the background, the lowest possible knee location should be the easiest to walk on. However, the look of the prosthesis is out of the ordinary because of the asymmetry. This is justified since prosthetic devices have always been a unique field of design and the potential benefit to gait may outweigh the decrease in appearance. The big picture for this type of prosthesis is that an inexpensive system such as this can provide a comfortable gait and can reduce the energy costs of the user.

The Background Chapter of this thesis covers a range of topics. Human gait analysis is the basis for prosthetic development. The various phases of gait is explained followed by gait asymmetries that arise due to the use of transfemoral prostheses and due to stroke. Gait after stroke is an interesting observation because stroke introduces a damping effect in the knee joint that causes a lot of interesting gait changes similar to gait changes in prosthesis users. Prosthetic types and problems with current transfemoral prosthetic designs are also analyzed. This is important to get a good understanding of the knee, ankle mechanisms, and type of compensation strategies amputees employ to have a stable gait. Passive dynamic walkers have been successfully used to model human gait. This shows the importance of the computational model that asymmetries in the knee height can still maintain gait symmetry.

The prosthetic simulator design is elaborated in the Design Chapter. The chapter describes the designs of the prosthetic knee, ankle, thigh, shank, knee brace, and the safety analysis conducted on the assemblies. The knee is the most interesting part of the system since it is completely passive utilizing a weight/position activated locking mechanism. The foot design is also unique because it uses a kinetic shape instead of an ankle joint. This saves weight and makes it easy for manufacturing. The thigh and shank are completely adjustable at 20mm differences. The thigh and shank are connected to the knee and allows the shifting of the knee lower or higher. The final part of the Design Chapter deals with the safety of the design. SolidWorks assemblies are analyzed with simulated ground reaction forces. The forces were deliberately chosen at the higher spectrum to ensure that the design is safe for any form of impact and cyclic loads.

The chapter on research results demonstrates the efficacy of all the claims. The data which provides the basis to prove that the system improves energy costs for the user is represented in the form of graphs for easy interpretation. The tests were conducted on the VICON system using 8 infrared cameras and markers to obtain precise motion capture data. Three test subjects were used for the experiment. The normal walking data of the participants is captured and it is then compared to the data when the participants were walking with the prosthetic simulator. Every participant walked on the prosthesis with the knee location at three different heights high, medium, and low depending on the participant's height.

CHAPTER 2: BACKGROUND

2.1 Human Gait Overview

Human gait is a cyclic walking pattern created by putting one leg in front of the other to move forward with a leg trajectory from back to front. During healthy human gait, legs move in a symmetric fashion that is always 180 degrees out of phase. Gait is a result of various complex movements occurring in synchronization. Humans start walking from a very early age where they develop the neural networking to activate specific groups of muscles in the legs that help in walking [16]. A complete gait cycle involves two major phases that each leg follows. In turn, each phase is subdivided into four subphases.

2.1.1 Phases of Human Gait

The two major phases of human gait are the stance and swing phase [67] [92]; these phases are depicted in Figure 2.1. The stance phase is the time the foot is in contact with the ground, whereas the swing phase consists of the time when the foot is swinging in the air. When one leg is going through the stance phase the other leg goes through the swing phase. During walking, there also exists a phase called double support where both the feet are in contact with the ground, which accounts for approximately 10% of the gait cycle. When the heel strike occurs on the stance leg, the swing leg is toeing off during its pre-swing phase. During the loading response sub-phase, the weight begins to shift to the stance leg meanwhile the swing leg initiates swing. During the mid-stance subphase,

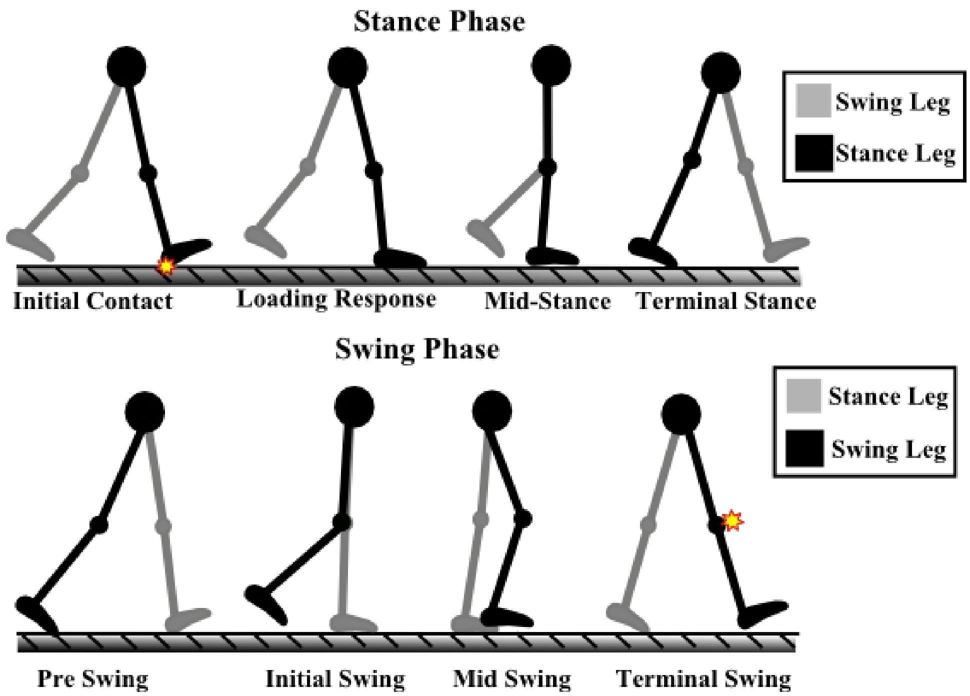


Figure 2.1: The 8 Phases of Human Gait. The Stars Represent Heel Strike at Initial Contact and Knee Strike at Terminal Swing.

all the load is shifted to the stance leg and the swing leg is in mid-swing. During the terminal stance subphase, the stance leg prepares for toe off and the swing leg initiates heel strike.

2.1.2 Transfemoral Amputation

Transfemoral amputation is the amputation of the leg above the knee. This type of amputation is usually performed as a result of trauma, accidents, or due to disease, like diabetes, vascular disease etc. With this type of amputation a person loses two of the most versatile joints in the human body, knee and ankle. The knee joint is important to human gait because it serves as a junction for the thigh and shank muscles. The knee locks and unlocks during heel strike and toe off respectively. Knee locking can be caused either by contraction of muscles (voluntary) or a slight overextension of the knee (involuntary). Without locking of the knee, human legs would buckle and walking would not be possible. Ankle joints are important in gait because the joint offers stiffness to avoid collapse of the leg at dorsiflexion or heel strike; at planarfexion or toe off it provides control and power to propel the body forward. The loss of function of these muscles results in variation of gait, usually as age progresses [45].

2.1.3 Gait Changes with Transfemoral Prosthesis

Transfemoral prostheses are designed specifically for people who have undergone transfemoral amputation. The prosthetic leg consists of a socket for the residual limb, a prosthetic knee and a prosthetic ankle. All prostheses to date aim to have the prosthetic knee in a symmetric position matching the intact leg. The prosthetic knees and ankles for various models of prostheses available are designed to mimic natural size, weight, and motion as close as possible. Amputees fitted with such a prosthesis tend to compensate with their stride length rather than step rate [43]. This means that the amputee spends more time on their biological leg than their prosthesis [38]. Jaeger et al. [43] noticed that the amount of asymmetry in gait has a relation with the stump length; the longer the stump length the lower the asymmetry. Knee disarticulation [4] [7] is a type of amputation that leaves the femur and the patella intact and it is more advantageous than transfemoral amputation because the amputee has all the thigh muscles intact. Abnormal spatio-temporal distribution of center of pressure was identified as another reason for asymmetry in transfemoral amputee gait [80]. Spatio-temporal parameters include the measurement of step length, step width, walking speed and cycle time. Center of pressure of the prosthesis fails to shift towards the posterior during gait initiation [93] and anterior during gait termination [94], whereas ideally it should.

Unilateral transfemoral amputees expend more energy than unilateral transtibial amputees (below knee) and healthy subjects [41][6]. This shows that there is a correlation between the energy expenditure and number of joints lost. This functional loss is because of the missing joints, or degrees of freedom, in the amputated leg. To equalize the functional losses, the body has to work harder; in this case it is the intact leg that experiences an increase in joint force moments and has to expend higher energy [63]. There are also residual stresses that are experienced in the stump as well, resulting in discomfort while walking [77]. The stresses in the residual stump are a bi-product of asymmetric reaction forces and moments. A study conducted by Mattes et al. [54] showed that as the mass and moment of inertia of the prosthesis approaches that of the intact leg the more asymmetric the gait becomes, therefore, expending more energy. Alignment of the prosthesis is also identified as a factor to increase energy expenditure [79]. The asymmetries in alignment also give rise to postural asymmetries. Gaunaurd et al. [18], showed that leg length, pelvic inclination, and hip extension were different for the intact and the amputated leg. Rabufetti et al. [69] showed that transfemoral amputees adapt asymmetric pelvic tilt and joint angles at the hip as compensation strategies during gait. The goal of the asymmetric prosthesis is to alleviate any form of compensatory motions, which in-turn reduce the misalignments of the prosthetic joints during walking. All transfemoral prosthesis designs so far have the knee location matching the intact leg. To mitigate gait asymmetries, I am investigating the use of an asymmetric transfemoral prosthesis. There was no mention in literature of an asymmetric prosthesis before the study by Sushko et al. [88]. The asymmetric prosthesis is a simple and unique passive solution relying purely on tuning walking dynamics.

2.1.4 Gait Disorders Caused Due to Stroke

Stroke is caused by a sudden loss of blood to the brain, clinically known as a cerebrovascular accident. Stroke can cause multiple problems depending on the part of the brain that is affected. The side-effects can be total muscular dysfunction of limbs on one side of the body, an inability to speak, or understand speech, and blindness in one side. Stroke victims who suffer from damage in their central nervous system, specifically hemiplegia which is a paralysis of the arm, leg and trunk on the same side of the body, develop an asymmetric gait [48] [60] [75]. Stroke victims develop gait asymmetries which is comparable to transfemoral amputees. The asymmetry is mostly caused by the stroke affected leg lagging behind the healthy leg. The stroke-affected leg lags because of decreased strength, or the inability to generate voluntary contractions in the required muscle groups, and inappropriate timing of muscle activity [65]. This affects the individual's healthy leg as well

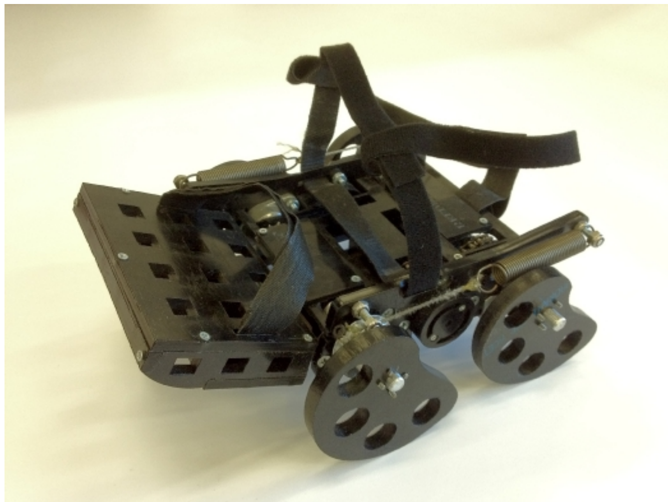


Figure 2.2: Gait Enhancing Mobile Shoes

since it has to expend more energy in compensating [13] for the lagging leg similar to transfemoral amputee gait.

Stroke victims, unlike amputees, have their limbs intact, therefore, rehabilitation of the damaged central nervous system of the affected side is possible. It is a matter of retraining the stroke-affected nervous system and muscles to restore the symmetric gait [33] [51]. Split-belt treadmills are employed to rehabilitate asymmetric hemiparetic gait. The treadmills have asymmetric belt velocities so the stride velocity of the lagging leg can be matched to the healthy leg [74]. However, the restoration of gait using split-belt treadmills is temporary in stroke victims since they were incapable to neurologically store the feed-forward walking pattern once they started walking on the ground [14] [73]. Other popular forms of rehabilitation are for a physical therapist to move the stroke patients limbs manually. This form of rehabilitation has now been automated by the use of exoskeletons [1] [3] [98].

The gait enhancing mobile shoes (GEMS) [14] [24] [29] is a recent development in stroke rehabilitation, shown in Figure 2.2. The GEMS is specifically designed to restore the gait permanently to the patient. GEMS performs two forms of rehabilitation: exaggeration and compensation [24] [25] [30]. Rehabilitation based on exaggeration is when the GEMS is worn on the lagging leg, the lagging leg is pushed backwards and motivates the user to perform a healthy toe-off.

Rehabilitation based on compensation is when the GEMS is worn on the healthy leg, this evens out the forward advancement between the healthy and lagging leg resulting in a symmetric gait. GEMS essentially reproduces the effects of a split-belt tread-mill over ground. The advantage of GEMS is that users can wear them over a long period of time performing ambulatory functions at home, which is important because retention is better if more time is spent in rehabilitation. Therefore, GEMS provides low cost stroke rehabilitation therapy that enables frequent and effective training at home [72].

2.2 Prosthetics

A prosthesis substitutes for or supplements the missing or defective body part. A prosthesis can be fitted externally or implanted into a person. Transfemoral amputation requires an external prostheses. Traditionally unilateral transfemoral prosthesis have always been made symmetric to the intact leg. This thesis is about development of an asymmetric lower limb prosthesis for transfemoral amputees where the knee is in an altered position relative to the intact knee, yet while retaining a symmetric gait. This section is intended to provide a comprehensive analysis of lower limb prosthetics starting with prosthetic development over the years, enumerate types of lower limb prosthesis and finally express the problems prosthesis users deal with.

2.2.1 History of Prosthetics

The purpose of this section is to have a brief review of the history of lower limb prosthetic devices. Prostheses have been in existence from the time of ancient Egyptians [89]. Early prostheses were simple stubs of wood that the person used to balance while walking. They did not assume an anthromorphic (human like) design until the late 18th century [64]. The prostheses before the 20th century were mostly passive. Due to the increase in the number of war related amputations, research began to develop better prosthetic devices. The first powered prosthesis was a pneumatic hand in Germany [10] and since then powered prosthetic designs have improved tremendously. Modern day prostheses feature microprocessor control and sophisticated hardware and software that mimic natural human gait. The high-end prosthesis technology is not accessible to everyone and still millions of amputees use semi-automated or passive prostheses. Now with the advancement in technology brings the possibility of inexpensive 3D printed prostheses [84].

2.2.2 Types of Prosthetic Knees

There are currently five general types of prosthetic knees available.

1. Manual Locking Knee
2. Polycentric Knee
3. Weight Activated Stance Control Knee
4. Single Axis Knee With Constant Friction
5. Knee with Outside Hinges

The manual locking knee is the most stable knee of the five but it has the lowest voluntary control; on the other hand, the knee with outside hinges is the least stable of the designs but has the high voluntary control. Control refers to the influence of the wearer on the device and stability is the influence of the device on the wearer. The Medicare Functional Modifier System [82] defines the "K" scale/score system to classify knees ranging from K0 - K4, where K0 is healthy humans and K4 is for children, bilateral amputees, and athletes. Most transfemoral amputees fall under K2 and K3 where they have to perform tasks in various environments.

The types of passive prosthetic knee joints can be further classified based on the design aspects. The design aspects are axes, friction, adjustability, mechanisms for locking, microprocessor control, and extension aids [34] [35] [37]. Axes that are commonly used are single, polycentric and exoskeletal. Common frictional components can be fluid based, pneumatic or hydraulic, or sliding based, constant, or variable. Locking mechanisms are manual, weight activated, and geometric lock. Extension aids are either internal or external. A combination of these design aspects coupled with

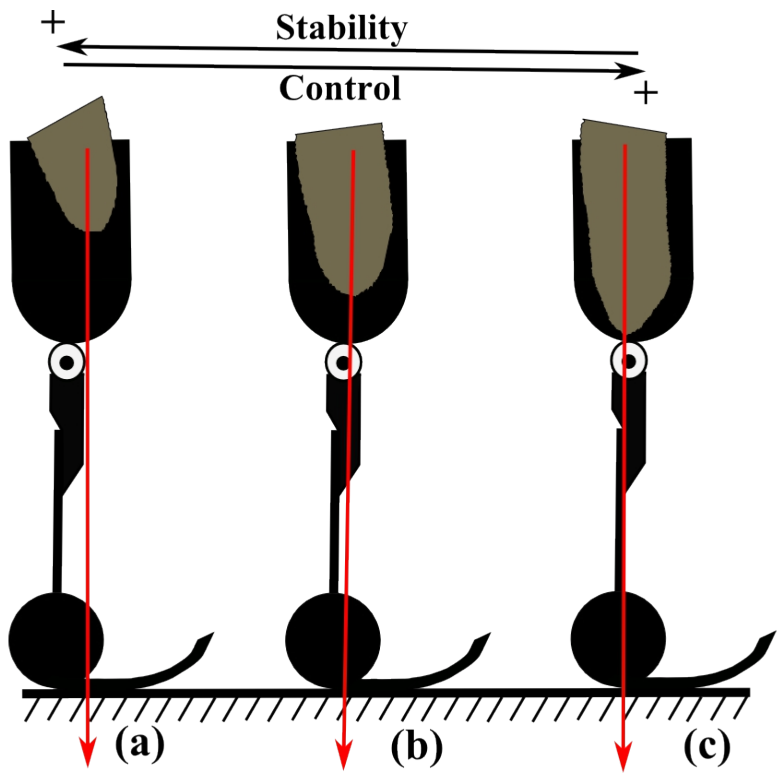


Figure 2.3: Design Change Based on Stump Length. (a) Short Stump with High Stability and Low Voluntary Control. The Thigh-Knee-Ankle (TKA) Weight Line is in the Anterior to the Knee Joint (b) Medium Stump with Medium Stability and Medium Voluntary Control. The TKA Weight Line is in the Middle to the Knee Joint (c) Long Stump (Knee Disarticulation) with Low Stability and High Voluntary Control. The TKA Weight Line is in the Posterior to the Knee Joint

the alignment of the knee's center of rotation relative to the Thigh-Knee-Ankle (TKA) weight line brings about the different types of prosthetic knee designs, design aspects shown in Figure 2.4. The design analysis is important since the asymmetric prosthetic design shifts the center of rotation axis below the anatomical center of rotation. Stability and control of a prosthesis can be explained with respect to TKA weight line, stability is high and voluntary control is low when the TKA weight line is anterior to the knee axis and vice versa when the TKA weight line is posterior to the knee axis, shown in Figure 2.3. The type of knee can further be related to the stump length of the amputee, long stump length, typically knee disarticulation [4], requires a low stability high control prosthetic knee joints because the amputee has enough residual muscles to have voluntary control and can maintain stability. Short stump lengths need high stability and low control prosthetic knee joints, because the amputees do not have enough muscles to have voluntary control and maintain stability.

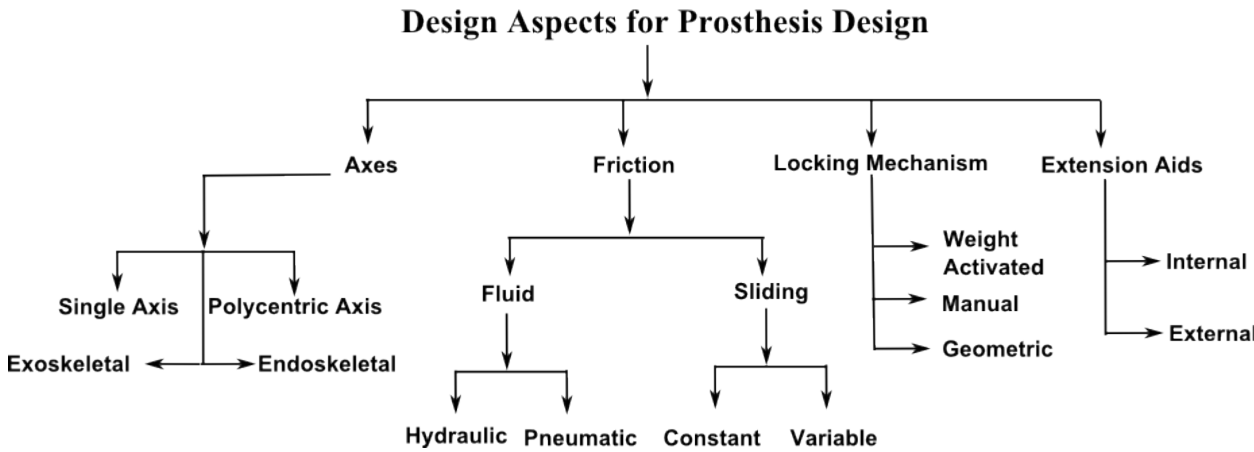


Figure 2.4: Design Aspects for Prosthesis Design

2.2.3 Active Prosthetics

This section is a brief overview about active prostheses, discussing their working, advantages, and disadvantages. Active knee and ankle prostheses are designed specifically to mirror the motion of the intact leg while walking. Active prostheses are microprocessor controlled and they use sensors to track the position of the knee and ankle joint, applied forces, stride lengths, stride velocity, and other measurements in order to acclimate to variation in speeds, activities, and environments [34] [36]. Actuators used to power these prosthetic devices are fluid based, pneumatic or hydraulic actuators, or electric motors. Pneumatic prosthesis designs have precise control and have enough power to assist in positive work, pneumatic prosthesis have demonstrated effective traversal over level surfaces [85]. A finite state based impedance approach was taken to control the prosthesis during walking and standing [86] [99]. Hydraulic systems are heavier than pneumatics, but offer higher power to weight ratio with lower leakages. A study by Yokogushi et al. [97] showed a successful design of a hydraulic polycentric knee that was capable of performing under various step rates and the results were similar to healthy human knee behaviour. Series elastic actuators utilize electric motors, in a study conducted by Martinez et al. [53] a prosthetic knee made out of series elastic actuators was modeled that was in agreement with normal knee mechanics. Active ankle prosthesis vary stiffness based on the activity performed by the user, microprocessor control [5] and electromyography control [2] have been employed in active ankle prosthesis.

Active prostheses adapt to the wearer's motion compensating for the lack of muscle control in the residual limb. Transfemoral amputees employing active microprocessor controlled prosthesss have shown significant improvements in gait and balance when compared to passive prostheses [44] [46]. This control provided by the active prosthetic systems resulted in lower energy expenditure when compared to that of a passive knee prosthesis [68] [81]. The only disadvantage to these otherwise proficient systems is their high price, need for on board power, and noise caused by actuators.

2.2.4 Passive Prosthetics

The Jaipur knee is a good example of a low cost prosthesis. Costing less than 40 dollars, it is manufactured by one of the world's largest prosthetic manufacturers in the world: the Jaipur-foot organization, India [61]. Passive prostheses rely on energy storage and release during gait [90]. Energy is stored during heel strike and it is released during toe off, in passive systems energy storage and release which is usually accomplished by springs. High energy expenditure during walking is one of the major reason amputees are unable to use them to the extent that they need [11]. The energy expenditure can be mitigated with good prosthetic design. Manual locking, polycentric, and weight activated stance control knees are popular when adapted for passive systems. This is because these prosthetic knee designs allow for simple mechanical and geometric locking that can function without microprocessor control.

The conceptual design by Sushko et al. [88] showed that an asymmetric transfemoral prosthesis was lighter than conventional prostheses, and can successfully restore symmetric gait. This thesis project utilizes a single axis knee with a weight/position translational locking knee mechanism and a passive ankle, using the design aspects described in section 2.2.2.

2.2.5 Problems Experienced by Transfemoral Prosthesis Users

Transfemoral amputees experience a myriad of problems affecting their quality of life [23]. Gait asymmetry is one among them discussed in section 2.1.2. Residual stump of transfemoral amputees is always enclosed in a socket, which is part of the prosthesis. In the socket the residual stump is affected by sweat and friction. This results in skin sores and other dermatological complications [15]. This section contains a summary of passive knee and ankle systems. Passive prostheses can either be internal, artificial knee implants [19], or external prostheses. This thesis focuses on transfemoral prosthesis and therefore this section is dedicated to examine external passive prosthetics. As discussed in section 2.2.3 the major disadvantages with the active prostheses is that they are expensive, need a portable power source, and the noise caused by the actuators. There are more amputees in developing countries than there are in all of the western world combined. Amputees in developing countries do not have the same opportunities as offered in countries such as United States of America. Therefore, there is a need for cheaper prosthetic devices that can make them walk and enable them to be a productive part of their society [83]. Passive knee and ankle joints with simple designs and cost-effective manufacturing is the solution for the less fortunate amputees. The sockets have to be tight around the residual stump to avoid slipping, this causes additional discomfort to the amputee. A biomechanical study by Gottschalk et al. [21] revealed that the loss of important groups of muscles, used for abduction and adduction in transfemoral amputees caused a loss of function in the hips and that preserving these muscle groups improved the walking conditions for amputees. The study also showed that comfort level for the amputee is dependent on the amount of residual tissue.

2.3 Passive Dynamic Walker

Passive dynamic walkers are mechanical devices which can walk downhill on a slope with only the force of gravity acting upon them [55] [56]. Passive dynamic walkers can be used to model human walking [28]. Passive dynamic walkers do not have any cognition unlike humans and therefore are suitable to model the sole kinematics of the human gait, in the absence of human feedback. The computational models of passive dynamic walkers can be turned to exhibit normal or pathological gait by varying PDW mass and mass distribution parameters. This section explains the importance of passive dynamic walkers in the design and validation of the asymmetric prosthesis.

2.3.1 Comparison to Actuated Bipedal Walkers

Actuated walkers are essentially robots with legs. They are classified into bipedal, quadrupeds and hexapods [66] [71]. Quadrupeds are four legged, like Bigdog by Boston dynamics [70], and hexapods, RHex [78], are six legged robotic models. Bipedal robots are two legged robots, some are essentially actuated passive dynamic walkers [12] [96]. These bipedal robots have motors that power the hip, knee, and ankle joints. The power permits these bipedal robots to walk on level ground by activating the joints to initiate gait. Passive dynamic walkers on the other hand can only walk down a slope, using gravitational force to move the joints. Passive dynamic walkers rely on gravity to initiate and sustain motion. Actuated Bipedal walkers can exhibit natural gait, but they require quasi-static balance [59]. Quasi-static balance in bipedal robots is to keep them from falling forward or backward while shifting their mass from between the legs. McGeer [55] describes passive dynamic walker motion as controlled free-fall.

2.3.2 Comparison to Humanoid Walkers

Humanoid robots are bipedal and they have anthromorphic features. Honda Asimo, Petman (Boston dynamics), Atlas (Boston Dynamics), and NAO humanoid robot [47] are successful humanoid robots. These robots are different from passive dynamic walkers because of their walking pattern. Stable dynamic walking does not have any autonomy in navigation [9]. Humanoid robots are completely autonomous and have to be in control of their navigation at every stage. Humanoid robots maintain static equilibrium at every phase because it allows them to control their motion effectively. The increased need for control makes these robots less energy efficient compared to passive dynamic walkers. Passive dynamic walkers do not have the ability to do other tasks apart from maintaining a naturalistic gait [27]. Therefore, passive dynamic walkers are energy efficient walking systems that cannot perform other tasks, and humanoid robots can do multiple tasks but do not have a natural gait.

2.3.3 Significance of Passive Dynamic Walker

Passive dynamic walkers are significant to this thesis because they provide versatile models for analyzing human gait [28]. Computational models of passive dynamic walkers have helped us in the understanding the various charecteristics of human gait. This development led to the asymmetric passive dynamic walker model [39] [87]. This model was flexible to simulate multiple configuration of masses, limb lengths etc. Therefore, when a transfemoral prosthesis was designed using this model, it showed that the knee location can be shifted below the anatomical position [88]. These computational models are also useful to render and design bipedal robots [12] [96].

2.4 Evolution of Asymmetric Passive Dynamic Walker Model

The nine mass model was designed to model asymmetric gait [39] [87]. It is a derivation of the five mass model [8]. The evolution of the nine mass model is covered in this section, shown in Figure 2.5.

2.4.1 Rimless Wheel

Passive dynamic walker gait can be compared to a rimless wheel rolling down a slope [52], depicted in Figure 2.6. While rolling a rimless wheel is always stable as every spoke makes contact with the ground, and it maintains its dynamic equilibrium by rolling. This is essentially dynamically stable free fall. Rimless wheels can also be used as walking mechanisms for robots [42]. A recent

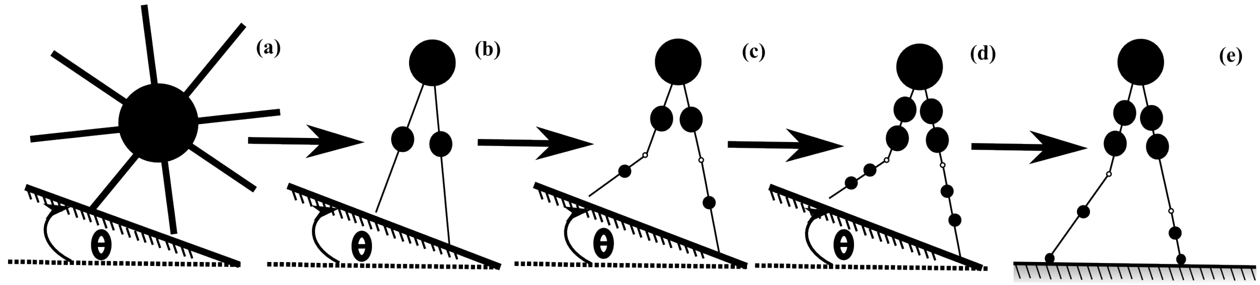


Figure 2.5: Evolution of Prosthesis. (a) Rimless Wheel (b) Compass Gait (c) Five Mass Model (d) Nine Mass Model (e) Asymmetric Nine Mass Model

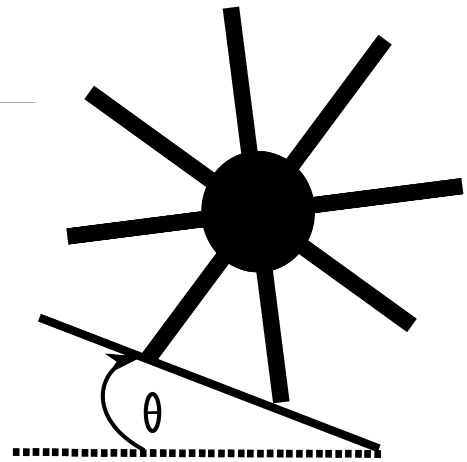


Figure 2.6: Rimless Wheel

study by Harata et al. [32] showed the design of an asymmetric rimless wheel with knee joints. The adjacent spokes of the asymmetric rimless wheel behave like a passive dynamic walker when it rolls.

2.4.2 Compass Gait

The first model of the passive dynamic walker was a straight leg model by McGeer [55], also known as the compass gait model. The compass gait walker model is equivalent to a double pendulum. It consists of three point masses, two of which are legs and the third is to simulate the hip joint [20]. This model is very limited since it does not have knee joints, and cannot be used to model normal human gait. However, hip moments and forces can be represented well in different gait conditions. This model was the predecessor to the passive dynamic walker with knees [57].

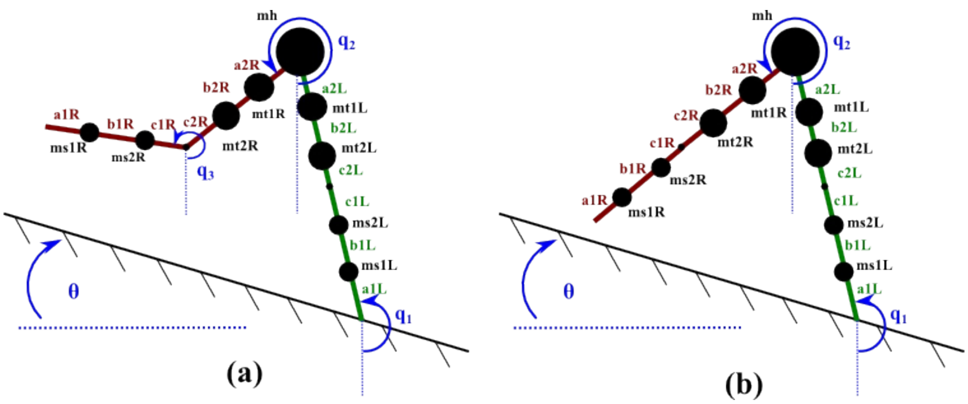


Figure 2.7: PDW Link Phases. (a) 3 Link Phase, (b) 2 Link Phase

2.4.3 Five Mass Model

The five mass model derived from Chen's [8] passive dynamic model was the step up from the compass gait model [55] and the PDW with knees [8] [57]. Five mass PDW models have two discrete phases of gait: 3-link phase, 2-link phase, depicted in Figure 2.7. For a normal gait cycle to exist, the energy lost during knee strike and heel strike has to be gained back from the effect of gravity on the overall inertia of the walker. This is similar to human gait where the energy stored at heel strike is released during toe off. The five mass model dynamics for various events were derived by Honeycutt et al. [39] and Sushko [87]. The five masses represent the hip, two thigh masses and two shank masses. Center of mass in the five mass walker can be changed to the anterior or posterior of the hip mass by changing the changing the moment of inertia [87]. The masses are not adjustable and assume the center of mass position of the hip, thigh and shank. This model was effective in presenting the mass distribution of a physical passive dynamic walker [40].

2.4.4 Nine Mass Model

The nine mass model is derived from the five mass model. The nine mass model is more versatile and adaptable to simulate mass distribution on the limbs. Four masses were added to the walker, one for each thigh and shank. The computational model of a passive walker utilizing the nine mass model was used by Honeycutt [40] to design and build a passive dynamic walker. The nine mass model was further developed by Sushko [87] where it was proven to be a better model than the five mass model to analyze asymmetric configurations. The center of mass and the moment of inertia can be varied independently in the nine mass walker unlike the five mass model.

2.4.5 Symmetric Passive Dynamic Walker Studies

Passive dynamic walkers so far, have been designed with symmetric knee location. This has allowed us to study and analyze healthy human gait [28]. There are several gait models that are available as discussed in section 2.3.2, but only the nine mass is relevant for the thesis. The nine mass model can accurately represent mass distribution, moment of inertia and other characteristics of human gait in a symmetric configuration. Honeycutt [40] used the nine mass model to design a 2 legged passive dynamic walker with symmetric gait. This was the proof of concept that the data from a computational model of the nine mass system can be used to design real systems.

2.4.6 Asymmetric Passive Dynamic Walker Studies

Analysis carried out by Sushko [87] on the nine mass model investigated asymmetric gait. The experiments were based on changing the moment of inertia on one leg and center of mass on the other. The changes gave rise to large asymmetries which then could be corrected by adjusting the center of mass or the moment of inertia. It is interesting to note that the configurations need not be the same to achieve symmetry. A recent study by Handzic et al. [62] showed that passive kinematic synchronization enables two dissimilar dynamic rotating systems to achieve the same motion without any sort of intra-system coupling or intervening control laws. Their model revealed that two double pendulums with different configurations of masses can swing in synchronization. This research is relevant because walker and human legs are systems similar to double and triple pendulums.

Therefore, if there is an asymmetry in their walking pattern it can be mitigated by adding masses to an appropriate configuration.

2.4.7 Prosthesis Design Based on Asymmetric Passive Dynamic Walker Theory

The adjustability of the nine mass model made it possible to achieve symmetric gait in a model with different leg lengths. Sushko [87] was able to demonstrate that a maximum of 68% removal of shank mass and a 13.4% removal of the total prosthesis mass can exhibit symmetric gait. This meant that the location of the prosthetic knee can be below the anatomical position. Sushko et al. [88] laid out the design of a conceptual asymmetric prosthesis with the knee location shifted below the normal position. Therefore, the addition or removal of masses and their positions can be computationally calculated for every person's geometry and will speed up the process in prosthetic design. The testing of this model will be conducted on normal human subjects fitted with a prosthetic simulator [85] [49] [91] [22]. The difference is that the knee location in the asymmetric prosthesis can be in the TKA weight line, because the knee location is below the natural knee line.

CHAPTER 3: DESIGN

3.1 Problem Statement

The goal of this research is to demonstrate that an asymmetric knee location can help improve the energy costs and symmetry in gait for transfemoral amputees. This thesis serves as a proof of concept. The prosthetic simulator has three different knee locations for every wearer. This design is specifically geared towards non-amputee wearers, for testing purposes. Testing it upon non amputees eliminates the compatibility problems that come with testing it on amputees, elaborated in Chapter 4. The idea behind the prosthesis is based on the computational prosthesis design derived from the nine mass passive dynamic walker [39] [87] [88], the concepts are discussed in detail in Chapter 2. This design aims to develop a prosthesis that is lighter than a human shank and has adjustable knee locations. The prosthesis can be used to analyze the variations in gait based on the knee location. As the knee location shifts downwards, the moment arm of the shank decreases, therefore, having a shorter shank swing phase.

This Chapter describes the design aspects of the asymmetric unilateral prosthetic simulator for use by non-amputees. The major components of this design are the passive knee, prosthetic thigh and shank, passive foot, and knee brace. The SolidWorks model of the complete assembly provides an accurate representation of the system, is depicted in Figure 3.1, and the complete system on a subject is shown in Figure 3.2 . Detailed SolidWorks diagrams with dimensions and photographs for the entire design can be found in Appendix A.

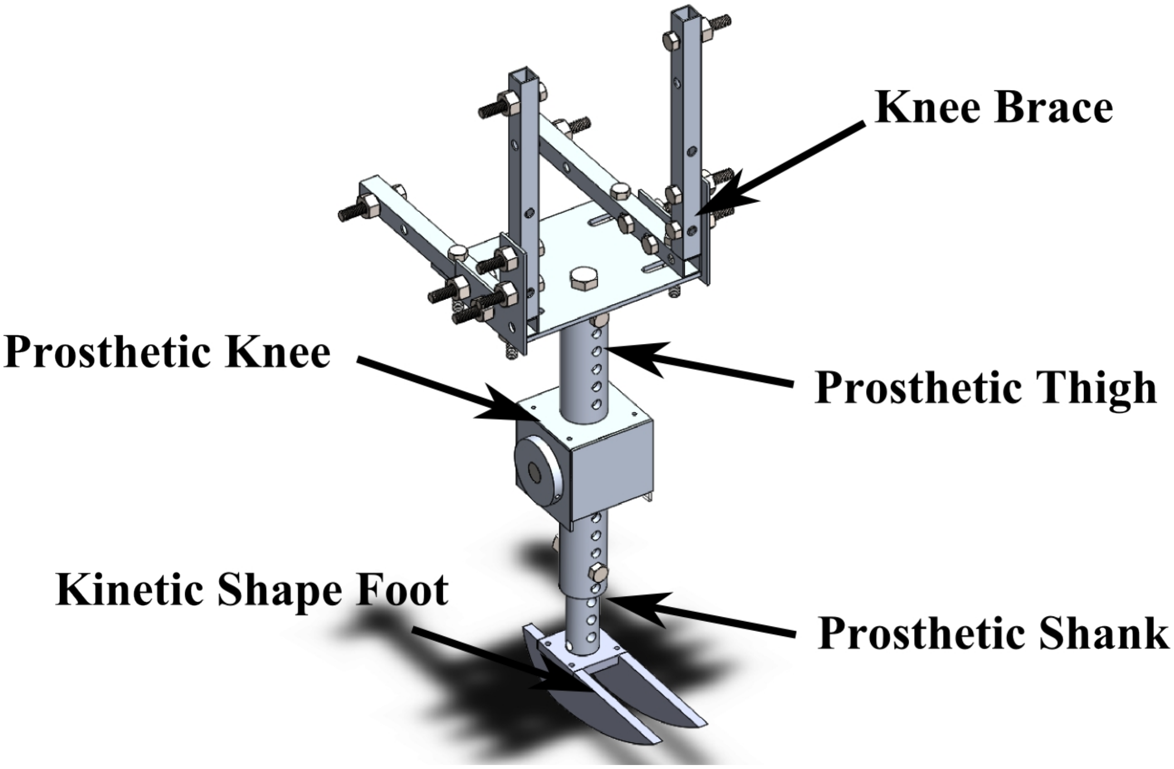


Figure 3.1: Complete Prosthetic Simulator Solidworks Assembly Model

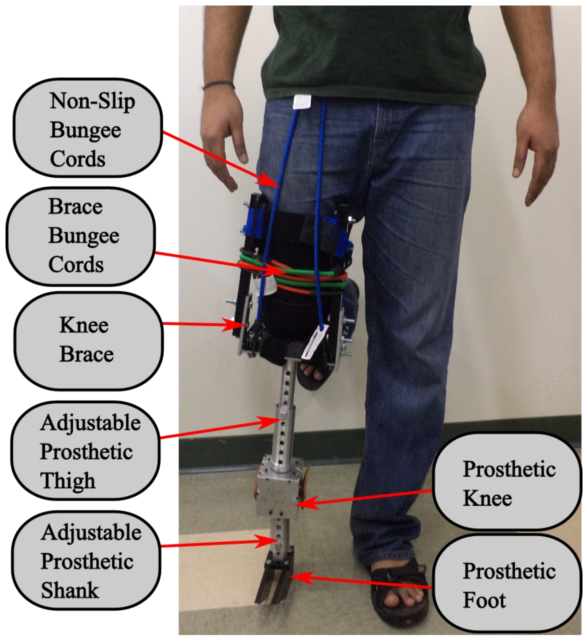


Figure 3.2: Prosthetic Simulator Fitted on Subject

3.2 Prosthetic Knee

3.2.1 Introduction

The problem statement requires that the prosthetic knee be a completely passive system. The inspiration for the design stemmed from the mechanism of the human knee. The knee is a very complex mechanism, consisting of a wide range of ligaments, tendons, muscles, cartilage, bones, and synovial fluid. For simplicity, only the movement of the bones are considered for determining the knee mechanism for locking and unlocking. The depiction in Figure 3.3 shows that the tibia and patella perform translational, as well as rotational motion, over the femoral surface. The cartilage acts as a guiding path for the patella. The locking and unlocking mechanism of the knee during normal walking is completely passive, utilizing only the dynamic forces of forward motion. The knee can be locked in other positions apart from the extension position by activating muscles to lock the knee in place. However, this design being completely passive will incorporate locking only when the knee is in the fully extended position, where the wearer's weight pushes the system down to lock. The novel knee mechanism experimented within this prosthesis design is a position/weight activated locking mechanism, explained in section 3.2.2. The detailed drawing with measurements of the prosthetic knee are listed in Appendix A.

3.2.2 Prosthetic Knee Mechanism

The knee was designed with the intent to incorporate translational and rotational motion in the mechanism. The position/weight activated locking mechanism simplifies the complex trajectory by allowing only two degrees of freedom, vertical translation, and single axis rotation to the complete system, depicted in Figure 3.4. The vertical translation is achieved by a simple vertical slot allowing the knee joint to have a small vertical displacement of 15mm, depicted in Figure 3.4. This translational freedom is utilized to lock and unlock the knee. The knee is locked when the weight of the

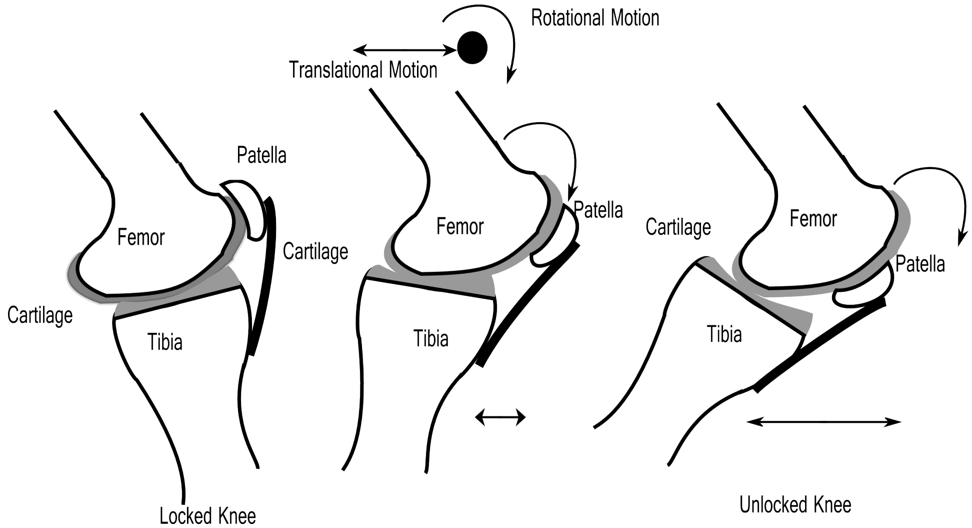


Figure 3.3: Human Knee Locking and Unlocking During Walking. Shows How the Tibia and Patella Rotate as Well as Translate Over the Femor.

wearer acts upon the knee assembly making the knee assembly reach its upper limit of its vertical translation. The locking is carried out by the half spur gear meshing with the gear rack fitted on top of the knee housing, depicted by Figure 3.5 (e). The knee starts to unlock as the wearer releases their weight at toe off and the force of gravity pulls the knee joint to the lower limit of its vertical translation, depicted in 3.5 (b). The single axis rotational freedom is provided to help the knee and shank perform a natural swing. The knee is free to rotate as its vertical position changes downward. Just before the terminal swing phase, the knee strikes the stopper to attain its position before the upward translation, depicted in Figure 3.5 (f). The gait with a prosthetic simulator worn by an able bodied person is shown in Figure 3.7 and the gait with an asymmetric prosthesis is shown in Figure 3.6.

3.2.3 Prosthetic Knee Specifications and Materials

The knee is required to overcome shock and transient loads generated during human walking. The maximum load on the prosthetic simulator can be as much as three times the wearer's bodyweight [50]. Assuming the wearer's weight to be a maximum of 100 Kg, the dynamic transient forces, which are higher than the heel strike force, is about three times the person's bodyweight.

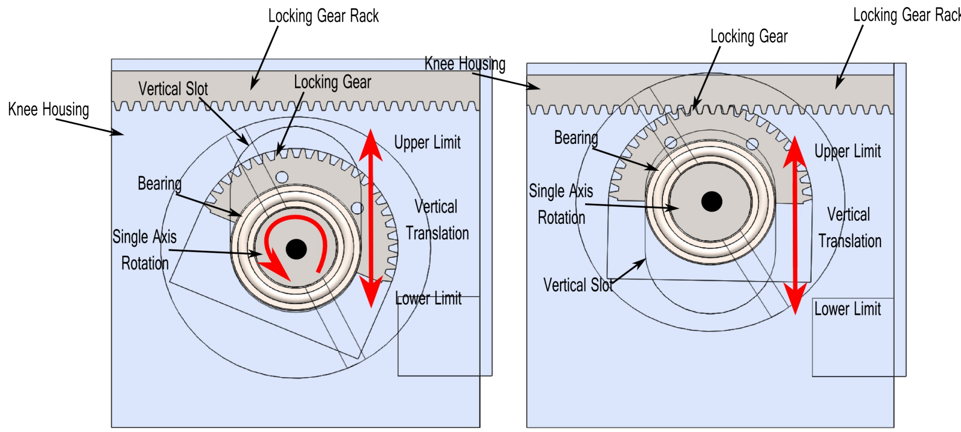


Figure 3.4: Simple Prosthetic Knee Mechanism. Two Degrees of Freedom, Vertical Translation and Single Axis Rotation

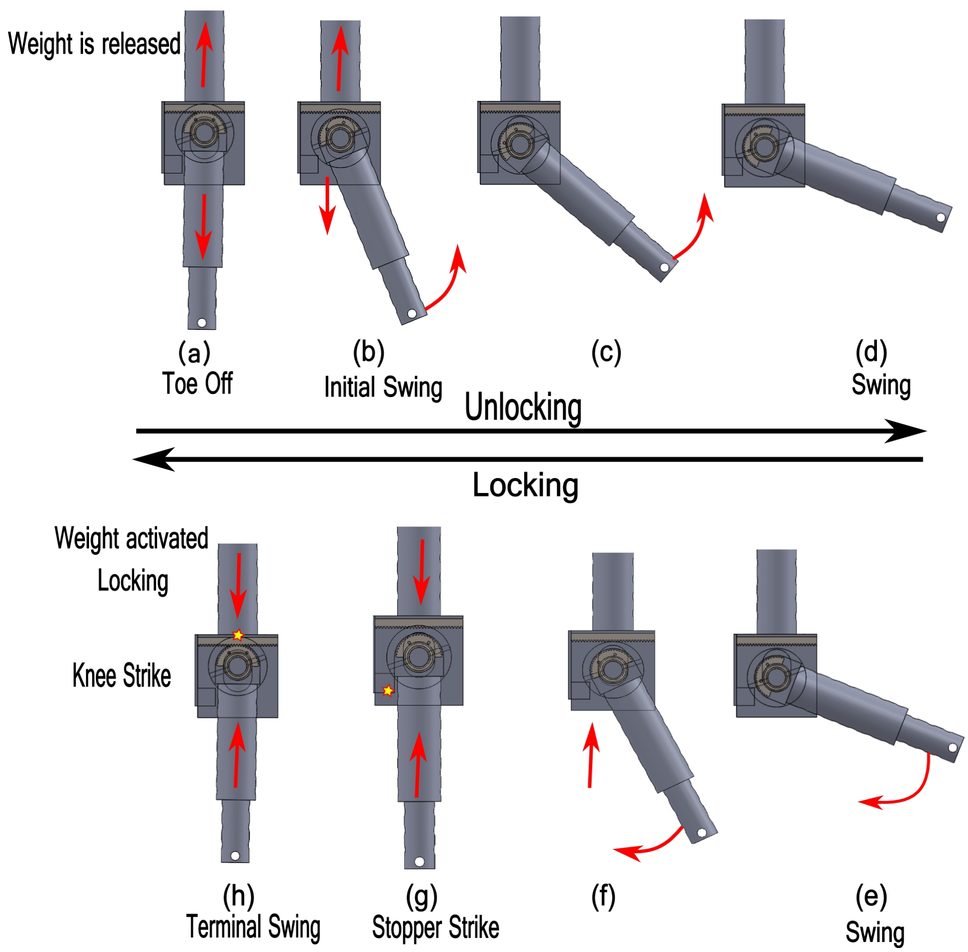


Figure 3.5: Mechanism of the Knee. SolidWorks Model Representing Knee Locking and Unlocking

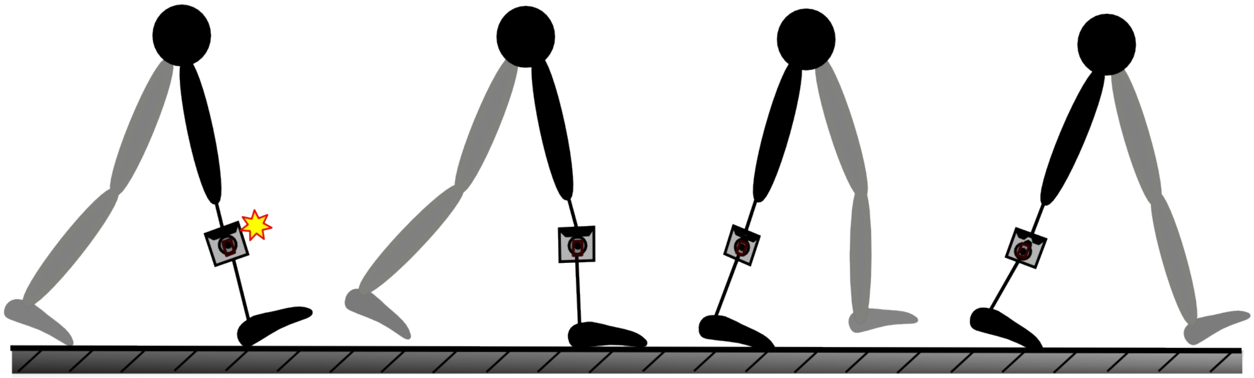


Figure 3.6: Gait with the Asymmetric Prosthesis.

Figure 3.7: Gait with Asymmetric Prosthetic Simulator

Therefore, every component of the knee should be able to withstand close to 3000N of force at any given point. The knee is designed to be effectively heavier than the prosthetic thigh and shank, as seen in Table 3.1. This ensures that the center of mass of the prosthesis is near the knee. This allows the center of mass to change based on the position of the knee in the prosthesis. The change in center of mass is an important observation provided by Susko et al. [88], which explains that a symmetric gait can be achieved by varying the center of mass of the knee to asymmetric locations with counter weights on the intact leg.

The knee is made out of a combination of aluminum and steel. Aluminum is used to build the knee housing, knee block, and collars. Aluminum is readily available, easy to machine, and has a high strength to weight ratio. However, steel has a higher shear strength and hence the gear, gear rack and shaft are made out of it. The shaft, gear, and gear rack experience shear and shock loads every gait cycle, aluminum will fatigue easily under such conditions.

3.3 Prosthetic Thigh and Shank

3.3.1 Purpose

The thigh and shank are simple adjustable linkages. They are used to adjust the height of the prosthesis to allow different knee locations. They are made out of aluminum cylinders, one hollow and one solid cylinder. The links are fit in place with the help of a bolt.

3.3.2 Settings

The thigh and shank are similar in design (consisting of two cylinders) with a maximum length of 240mm and minimum length of 140mm. The thigh and shank have six settings each allowing for 12 total settings. Each setting gives the user 20mm of difference in length, depicted in Figure 3.8. There is a constant length of the knee (Knee Top Plate + Gear Rack Width + Knee Block) and the foot height equivalent to 105mm. Therefore, depending on the anatomical shank length of the wearer, the settings can be decided keeping in mind the constant lengths. The detailed drawings of the prosthetic thigh and shank are depicted in Appendix A.

3.4 Foot

3.4.1 Foot Designs of PDWs

The foot designs of passive dynamic walkers offered an insight into passive foot designs that can be employed on the prosthetic simulator. Passive dynamic walkers have been modeled with a point foot, curved foot, and in more advanced biped walkers ankles that can provide forces similar to dorsiflexion. The point foot is an easy analytical model that was used by Chen [8] to analyze the five mass model. The curved foot model was first proposed by McGeer [56], that is a constant radius foot with a radius approximately one third leg length. This foot shape allowed the PDW's legs to clear

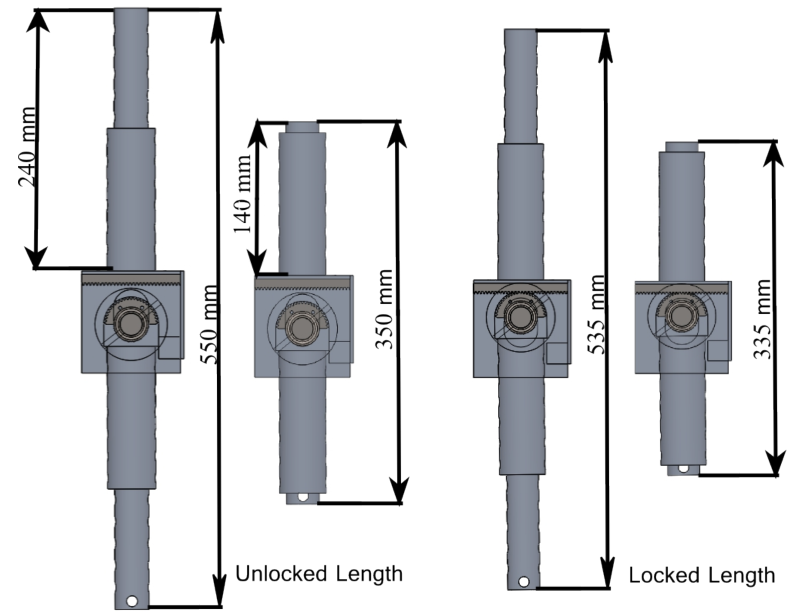


Figure 3.8: Prosthetic Thigh and Shank Length Settings

the ground easily. The foot design by Honeycutt explored the possibilities of testing changing radius foot designs in passive dynamic walkers, which would enable the foot to release the energy stored during heel strike at toe off [40]. In an another study they showed that constant radius foot designs can be replaced with flat foot designs that were mounted on the ankles using torsional springs [95]. In this project I chose to keep the foot design as simple as possible. The design resulted in a foot that does not require an ankle.

3.4.2 Rollover Shape

The foot design was based on roll over shapes. Roll over shapes are defined by the change in center of pressure of the foot during walking. There are three main phases when the center of pressure at the foot changes: stride initiation, steady state, and termination. In this design a constant radius roll over shape, which is one third of the total leg length, is considered [58]. The assumption is made based on the roll over shapes analyzed in [31] [58] [67], which show that most of the roll over happens in the anterior of the foot.

Two roll over shapes were tested for the foot roll over shape, depicted in Figure 3.11(a). The roll over shape for the first design has a constant radius which is one third of the leg length. This constant radius abruptly changes to a smaller constant radius near the posterior of the foot. This design failed testing because it tends to roll backwards towards the smaller radius, which made it difficult to walk forward.

The second foot design is based on the kinetic shape concept [26]. Kinetic shapes roll on flat surfaces when a force is applied on its axle. This foot shape will roll forward when a person applies their weight on it. This emulates dorsi-flexion of an ankle joint; the shape compensates for the lack of an ankle joint. The shape for this specific foot shape is shown in Figure 3.12. Equation 3.2 is the vertical force, which is based on the assumption that a person weighing 100kg will use the foot. Equation 3.3 is the horizontal force that will be generated when the force is applied. Equation 3.4 is the initial radius assumed for the shape based on one third leg length. These variables are substituted in Equation 3.1 which results in Equation 3.5. The foot design that was implemented is shown in Figure3.11(b), where the radius decreases towards the front of the foot. This design worked successfully and was implemented in the final design used for testing, shown in Figure 3.10.

$$R(\theta) = \exp\left[\int \frac{F_r(\theta)}{F_v(\theta)} d\theta + Constant\right] \tag{3.1}$$

$$F_v(\theta) = 1000N \tag{3.2}$$

$$F_r(\theta) = 50N \tag{3.3}$$

$$InitialRadius = 0.30m \tag{3.4}$$

$$R(\theta) = \exp\left[\int \frac{50N}{1000N} d\theta\right]_{\theta=0}^{\theta=\pi} + 0.30m \tag{3.5}$$

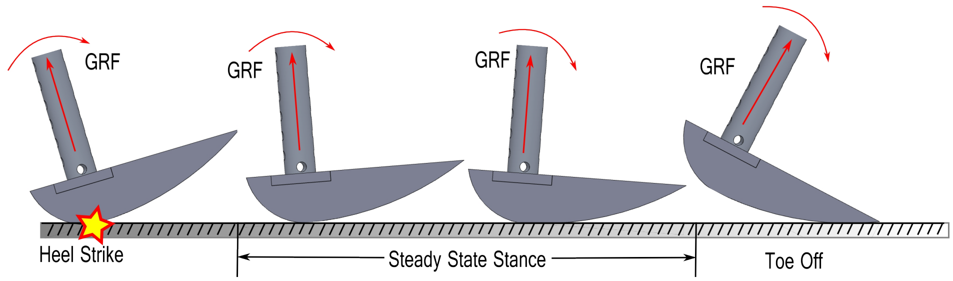


Figure 3.9: Mechanism of Foot Roll Over During Walking.

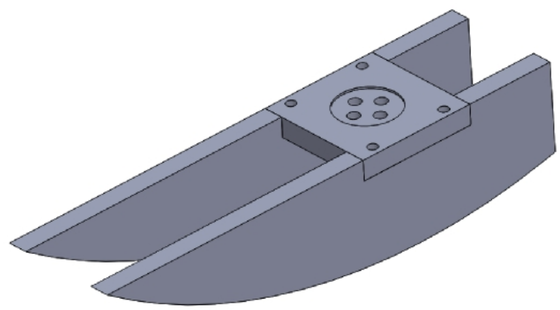


Figure 3.10: Foot Used for the Prosthesis

3.4.3 Mechanism

The foot assembly is fit rigidly to the lower solid cylinder. The foot does not need an ankle mechanism because of the kinetic shape rolls forward when the user applies their force on it. As the user's moment of inertia shifts forward, the foot starts to roll into a smooth forward motion leading to toe off, depicted in Figure 3.9.

3.5 Knee Brace

3.5.1 Introduction

This transfemoral prosthetic simulator is specifically designed for a non-amputee wearer. The design required an interface between the prosthesis and the wearer's leg. The wearer's leg will be

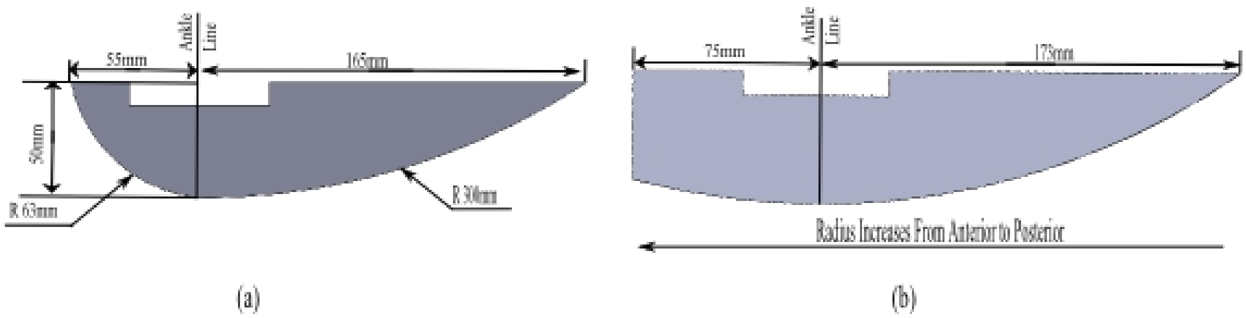


Figure 3.11: The Roll Over Shapes of the Prosthetic Foot. (a) It is the First Design with a Large Constant Radius Anterior to the Ankle Line and Smaller Radius to the Posterior (b) The Second Design which has a Constant Decreasing Radius Towards the Anterior and Constant Increasing Radius to the Posterior

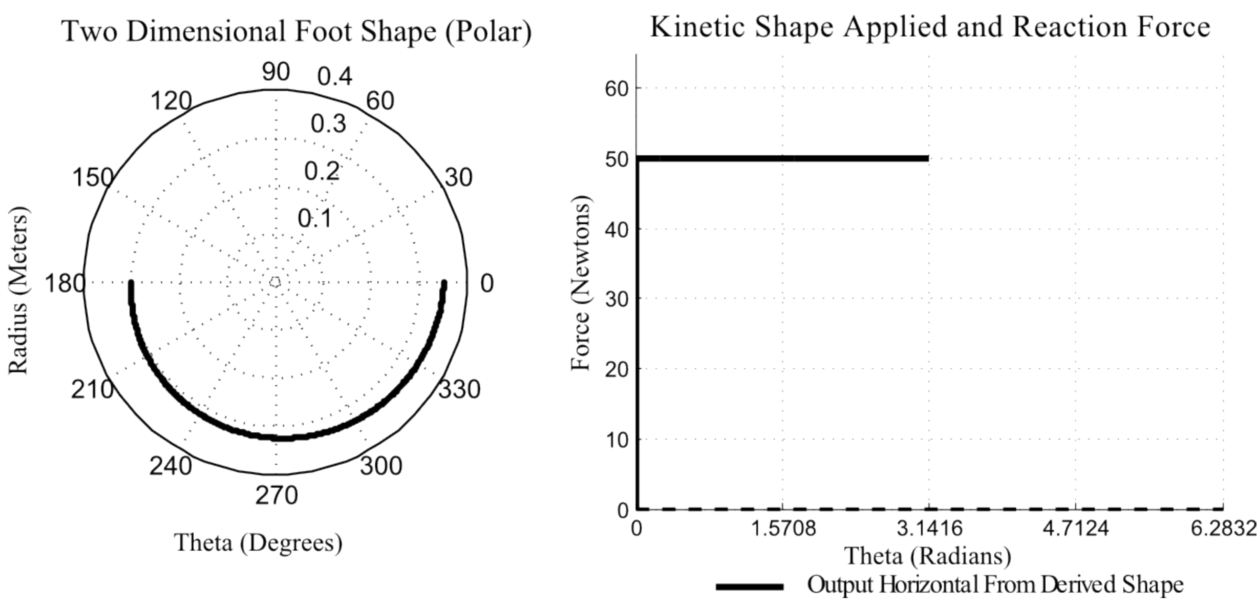


Figure 3.12: Kinetic Shape. (a) Foot Shape Polar Plot (b) Applied and Reaction Force Plot

held at a right angle and secured tightly. The prosthesis is designed to be fit on the knee brace by locking it on a bolt.

3.5.2 Requirements

The knee brace is required to be light weight, rigid, durable, and comfortable. This design is made completely of aluminum and the different components are secured with steel bolts, depicted in Figure 3.1. Aluminum is chosen because it is easy to machine and has a higher shear strength than delrin plastic and wood. The frame is rigid in order to restrain the movement of the wearer's leg because it can interfere with the motion of the prosthesis. Aluminum used for the brace is light and strong enough to withstand continuous load cycles. Table 3.2 shows the safety factors for the base plate which is the component that will experience the maximum load. The brace also defines the position of the prosthetic leg with respect to the Thigh-Knee-Ankle line, discussed in section 2.2.2. The prosthetic leg is placed to the anterior of the wearer's Thigh-Knee-Ankle line to ensure high control and low stability of the prosthesis. This is important because the wearer in this case has all the muscles intact in their leg. The positioning is subject to change when the prosthesis will be fitted upon amputees in the future.

Two unique problems came up while testing the knee brace. First, the knee brace tends to slip down. The only solution to this problem was to secure the brace to the safety harness using bungee cords, depicted in Figure 3.2. The second problem was comfort. The metal components were hard and would hurt the wearer while walking. To ensure the comfort of the wearer, they are required to wear flexible knee braces that provide enough padding. In addition to the padded flexible knee braces, the metal knee brace is provided with additional padding for comfort.

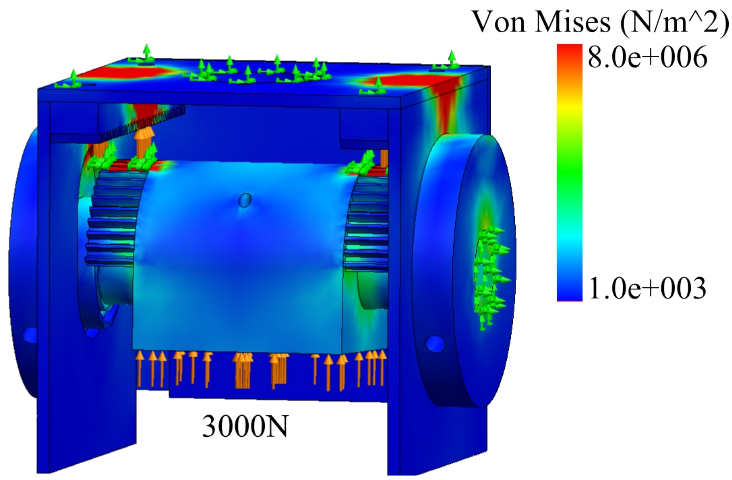
Future designs may look into a simple rigid frame with adjustable shank and thigh lengths to accommodate a wide range of users. The frame may also have an acute angle to have more clearance of the bent leg. The padding can be designed in such a way that it can arrest the user's leg movement.

3.6 Safety Analysis of Components

This section is to provide validation to the materials chosen and component design. All components are subjected to the maximum load of 3000N, assumed for three times the body weight of a person weighing 100kg [50]. Although the load may be shared by the components, the tests were to make sure that all components will not fail when maximum load is applied. The load simulations were carried out in SolidWorks Simulationxpress package. The prosthetic components were fixed at points according to their design constraints and a force of 3000N was applied on the components in the downward direction, for circular components it was applied on the surface. The testing was carried out for the aluminum and delrin components because they have a higher chance of failure than steel. This is because steel has a very high shear strength compared to aluminum and delrin. Table 3.2 shows the factor of safety of the components that are subject to the forces directly. Figures of component analysis can be seen in Appendix B.

Analysis was also conducted on assemblies. The main assemblies that were concentrated upon were the knee, foot, thigh, shank, and base plate. The knee assembly stress analysis depicted in Figure 3.13 shows that the top plate of the knee, ends of the shaft, and the locking gear are constrained, and a ground reaction force of 3000N is applied to the knee block and to the meshing portion of the gear rack. This results in high stress in the top plate, shaft and knee block due to high force acting upon them. The factor of safety is low because the knee block is made out of aluminum and it experiences a high shear force. The system on the whole, however, proves that the design was correct and will not fail under the given load.

The prosthetic thigh and shank analysis were performed under similar conditions as the knee. Figure 3.14 depicts the thigh and shank Von-Mises stresses. In the prosthetic thigh assembly, the top of the upper small cylinder is constrained and a force of 3000N is applied on the bottom of the upper large cylinder. This simulates the force coming from the top plate of the knee to the upper large cylinder. On the prosthetic shank assembly, the loading is the reverse of the thigh. As seen the



Minimum Factor of Safety for Complete Knee : 1.4
Minimum Factor of Safety for Knee Block : 3.2
Minimum Factor of Safety for Knee Top Plate : 1.4

Figure 3.13: Prosthetic Knee Analysis

maximum stresses are around the holes. The safety factor of the shank is higher because the force that acts upon the solid cylinder.

Analysis was performed on both foot designs, depicted in Figure 3.15. The top surface of the foot is constrained in order to avoid the calculation of maximum displacement, which is not important for this trial. A ground reaction force of 3000N is applied on both the designs and it is observed that the factor of safety of the second design is higher than the first. This change can be explained because of the shapes of the foot designs. The first design has two constant radii that change the roll over shape suddenly at the ankle line. In the second design, which has a constant decreasing radius, the roll over shape is gradual and is capable of distributing the force better than the constant radius design.

The base plate is analyzed using the thigh assembly attached. The attachment bolt hole is constrained and a ground reaction force of 3000N is applied on the bottom of the large cylinder. As seen in Figure 3.16 the hole experiences high stress concentration. The factor safety indicates that the design is still far from failing.

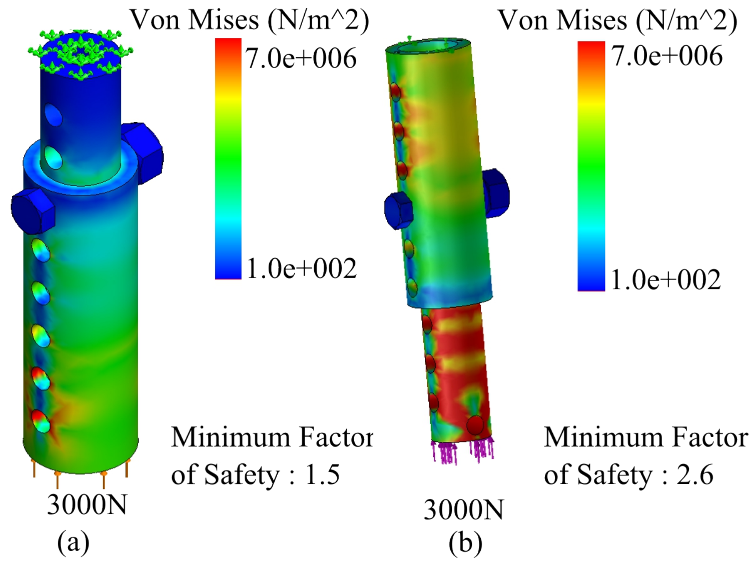


Figure 3.14: Prosthetic Thigh and Shank Stress Analysis. (a) Prosthetic Thigh Assembly with the Top of the Small Cylinder Constrained and Ground Reaction Force of 3000N Acting on the Bottom Circumference of the Large Cylinder (b) Prosthetic Shank Assembly with the Upper Circumference of the Large Cylinder Fixed and a Ground Reaction Force of 3000N Acting on the Bottom of the Small Cylinder

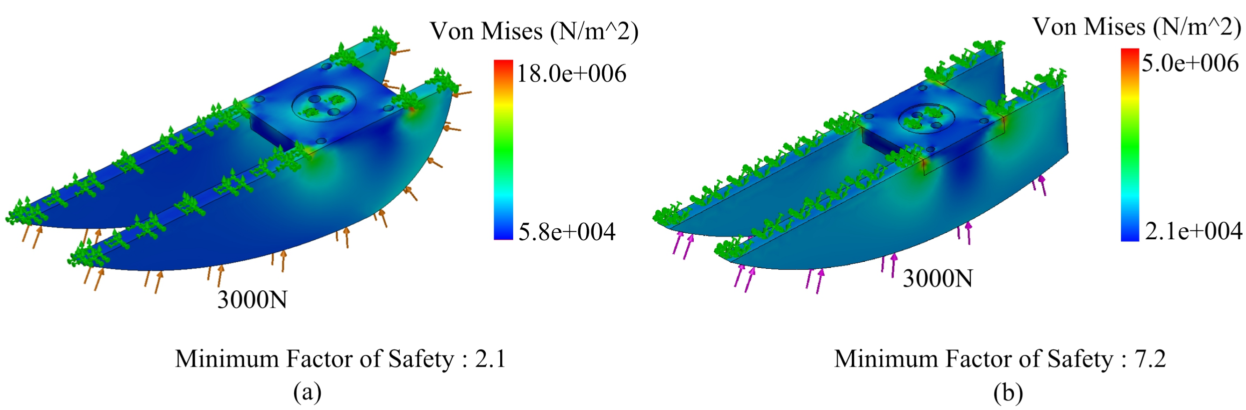


Figure 3.15: Prosthetic Foot Analysis. (a) Von Mises Stress Experienced by Constant Radius Design (b) Von Mises Stress Experienced by Shape Constant Decreasing Radius Design

Table 3.1: Mass of Prosthetic Components

S.No	Part	Weight (Grams)	Number of Components	Combined Weight (Grams)
1	Lower Small Cylinder	207.9	1	207.9
2	Upper Small Cylinder	223.8	1	223.8
3	Front Knee Plate + Stopper	129.1	1	129.1
4	Steel Gear Rack	38	2	76
5	Upper Large Cylinder	269.7	1	269.7
6	Knee Top Plate	90.3	1	90.3
7	Lower Large Cylinder	255.4	1	255.4
8	Knee Side Plate	80.6	2	161.2
9	Ball Bearing	60.5	2	121
10	Steel Shaft	241.4	1	241.4
11	Collar	89	2	178
12	Half Gear	87.4	2	174.8
13	Knee Block	276.5	1	276.5
14	Brass Connecting Bolt	33.1	1	33.1
15	Aluminum Bars	146	4	584
16	Line Holder	62.4	4	249.6
17	Right Angle Bracket	118.2	2	472.8
18	Base Plate	281.3	1	281.3
19	Bolt and Nut	26.3	22	578.6
	Weight of Knee Brace:		2.19 Kg	
	Weight of Extenders:		0.95 Kg	
	Weight of Knee:		1.5 Kg	
	Weight of Total system:		4.6 Kg	

Table 3.2: Miminum Factor of Safety for Prosthetic Components

S.No	Prosthetic Component	Miminum Factor of Safety
1	Prosthetic Knee	1.4
2	Prosthetic Thigh	1.5
3	Prosthetic Shank	2.6
4	Foot	7.2
5	Base Plate	1.4

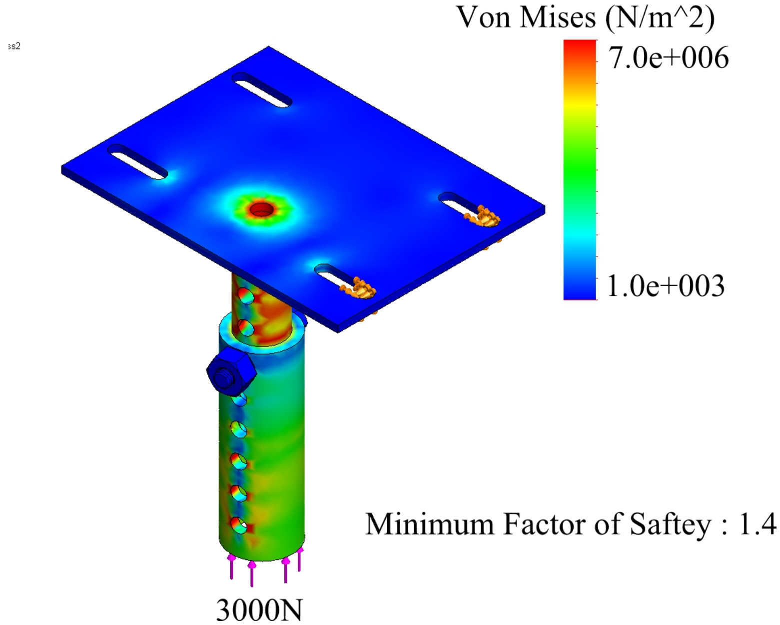


Figure 3.16: Base Plate Stress Analysis

CHAPTER 4: TESTING AND EXPERIMENTAL RESULTS

4.1 Kinetic and Kinematic Data Aquisition Protocol

The data acquisition was done in two stages. The first stage was to collect kinematic data on three subjects. The data was collected for trials consisting of normal walking, knee at high setting, knee at medium setting, and knee at lowest setting. A VICON system which has an accuracy of 1mm and sampling rate of 120 Hz was used to obtain the kinematic data. The VICON uses infrared cameras and reflective markers to accurately track every marker's motion in 3D space. The markers were put on the lower extremity of the participants as shown in Figure 4.1 for normal walking, and Figure 4.2 shows the marker placement with the prosthetic simulator.

The three participants chosen for the study were all male and did not have any gait disability. All participants wore the prosthesis on their right leg. The kinematic data was collected on all subjects for all four trials. Table 4.1 shows the height, weight, leg length, and shank length. The participants walked on a wooden platform, which is the same as overground walking. All participants followed an approved USF Internal Review Board (IRB) protocol.

Table 4.1: Participant Data

Subject	Weight (Kilograms)	Height (Centimeters)	Leg Length (Centimeters)	Shank Length (Centimeters)
1	96	186	98	43
2	85	189	110	55
3	108	184	98	52
Average	96.3	186.3	102	50
Standard Deviation	11.5	2.5	6.9	6.2

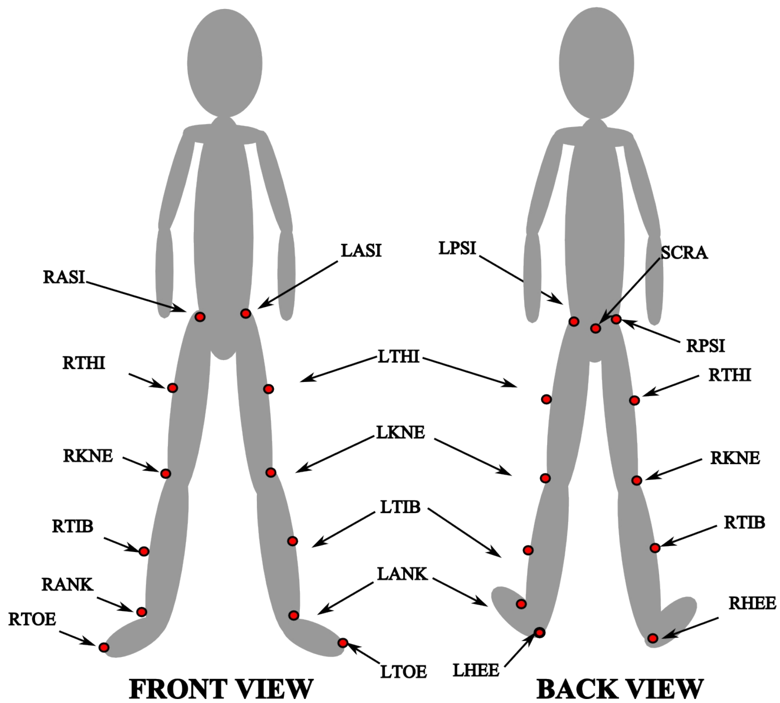


Figure 4.1: Full Body Marker Layout

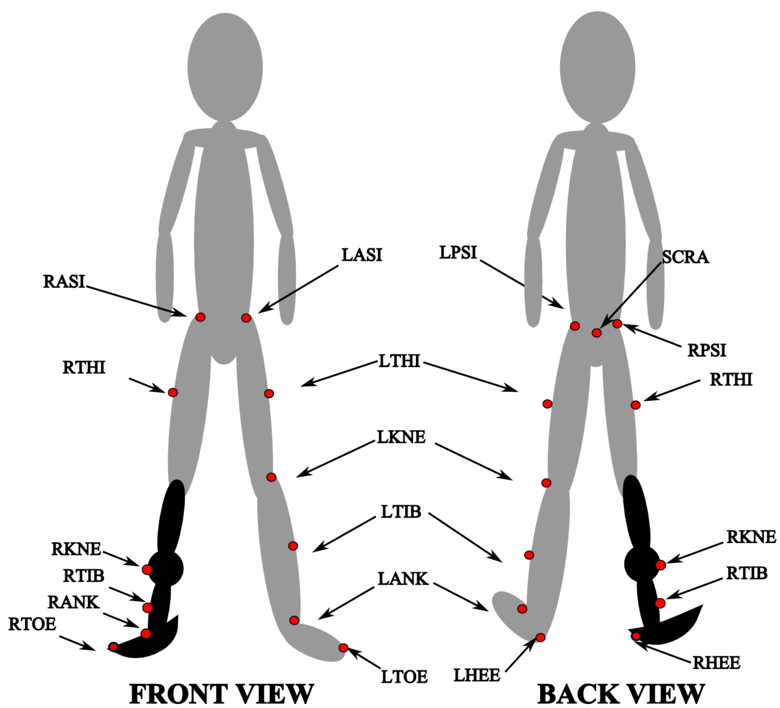


Figure 4.2: Markers for Prosthetic Simulator

The second stage was to collect kinetic and kinematic data. The trials for this data acquisition are the same as the first stage. A Computer Assisted Rehabilitation ENvironment (CAREN) system was used for this stage of testing, shown in Figure 4.9. The CAREN also has a VICON system for kinematic data acquisition. It also has a split-belt treadmill with force plates to measure the kinetic data. Therefore, the first stage is ground walking and the second stage involves treadmill walking.

Subject 3 from the first stage was put through the complete set of trials (normal walking, knee at high setting, knee at medium setting, and knee at lowest setting) for the second stage. The kinetic and kinematic data were recorded for both legs. The kinematic data for both stages were post processed to obtain useful data. The gait cycle of the skeletal frame from the processed data for the gait is shown Figure 4.3.

4.2 Kinematic Data Analysis for First Stage Testing

The step lengths were calculated by finding the difference between the position of the left heel and the right heel. Corresponding swing times were also found for each leg. Step length and swing time graphs were plotted for each trial, shown in Appendix C. A cumulative average and standard deviation was found for all subjects, shown in Figures 4.4 and Figure 4.5. Step lengths for all trials are symmetric. The step lengths of the knee at low setting are higher than the other settings. The medium setting is very close to the step lengths of the low setting. The knee at high setting is overall the farthest from normal walking step length. Therefore, the step lengths of the prosthesis at low setting are symmetric and closest to normal walking, which is the trend that was expected to be seen based on the passive dynamic model of walking.

The swing times were also found to be asymmetric. The left leg (healthy leg) has the same time for its swings for all cases. The leg with the prosthetic simulator has either a symmetric swing time as the healthy leg or takes more time to swing. Therefore, the cumulative step times do not show a pattern such as the one observed in step lengths.

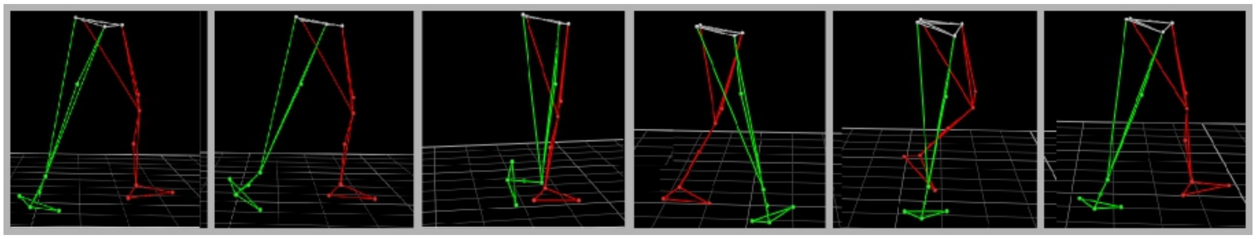


Figure 4.3: Gait with Prosthetic Simulator- VICON Model

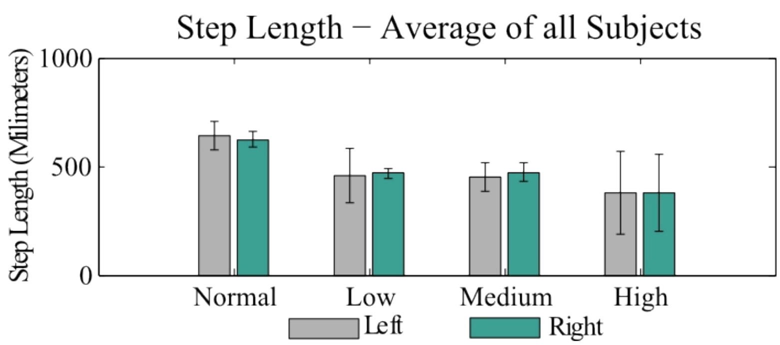


Figure 4.4: Cumulative Step Length Plot for All Subjects

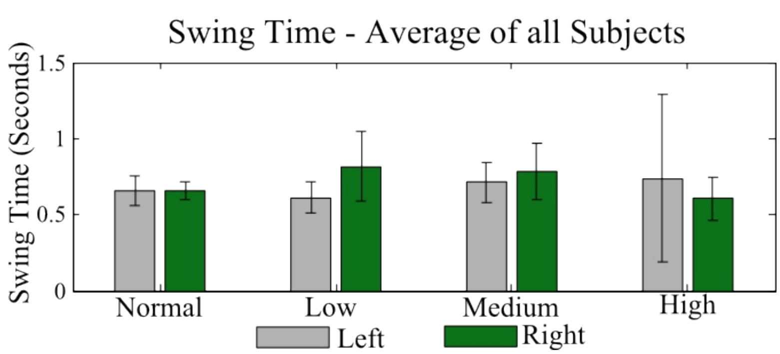


Figure 4.5: Cumulative Swing Time Plot for All Subjects

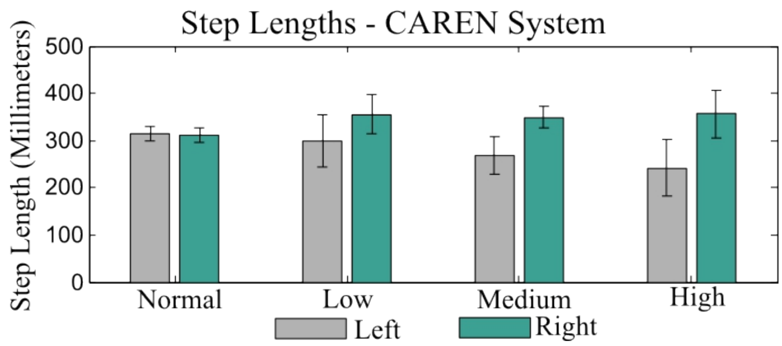


Figure 4.6: Step Length for CAREN System

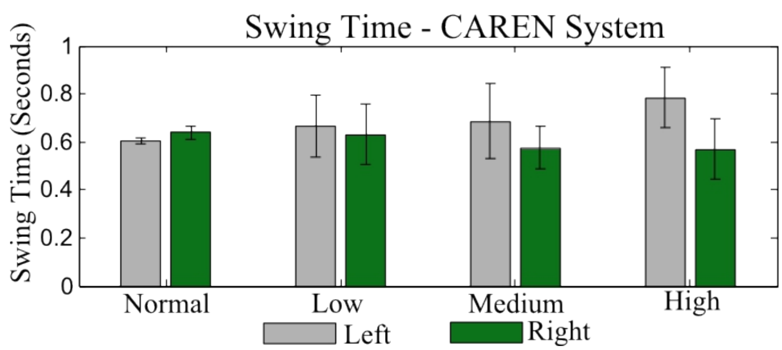


Figure 4.7: Swing Time for CAREN System

4.3 Kinematic and Kinetic Data Analysis for Second Stage Testing on CAREN System

Kinematic data obtained from the CAREN system is shown in Figure 4.6 and Figure 4.7. Looking at the step length graph it can be seen that the subject takes longer steps with the prosthetic leg; note that the swing time of the prosthesis is shorter than the left leg. Step lengths of the left leg are slightly asymmetric in the medium and the high knee setting. The low knee setting is the closest trial to emulate normal walking gait. The high knee setting shows the most asymmetry in step lengths. Swing times of the low knee setting is symmetric whereas the high knee setting is asymmetric.

Kinetic data of the ground reaction forces was also obtained for all the trials on the CAREN system, shown in Figure 4.8. Looking at the data it is apparent that the person applies more force on the prosthesis. This is because there is a high shock load that is applied on the prosthesis, this is another form of compensation mechanism. An interesting symmetry in the forces is seen in the high position; the subject was walking with slightly more force on the healthy leg than on the prosthesis.

The most significant trend observed is that the closest force that resembles normal walking is the trial with the low knee setting. More data has to be collected on different subjects walking to have a more comprehensive understanding of the kinetics of the prosthetic simulator.

4.4 Comparison of Data from Treadmill Walking and Over Ground Walking

The data acquired from the ground walking tests was small compared to the data acquired from the treadmill. This is because of the space constraints of a VICON system which cannot cover a very large distance for ground walking. This is not an issue with a treadmill; more steps can be recorded on a treadmill. Previous research on healthy subjects has not shown any statistically differences between overground walking and treadmill walking [76]. A recent study which compared overground walking with treadmill walking in the CAREN system of transtibial amputees and healthy subjects showed that overground and treadmill walking were similar enough, except for a slight variability in step width and step time results [17].

The previous sections analyze the data from ground walking and treadmill walking. The data for step length shows that over ground step lengths are more symmetric compared to the treadmill step lengths. The swing times for the left leg vary from ground walking to treadmill, but the prosthesis has almost the same times in both cases. A definite conclusion cannot be arrived to prove that there is a statistical significance given the small sample size especially for treadmill study which had only one participant. Further studies have to be conducted on more subjects to arrive at an exact conclusion for this comparison.

4.5 Discussion

The asymmetric transfemoral prosthetic simulator has its concept rooted in the passive dynamic walker model. A future expansion of this project is to compare the real world kinetics and kinematics of the prosthetic simulator to an asymmetric PDW model with the same characteristics of the device.

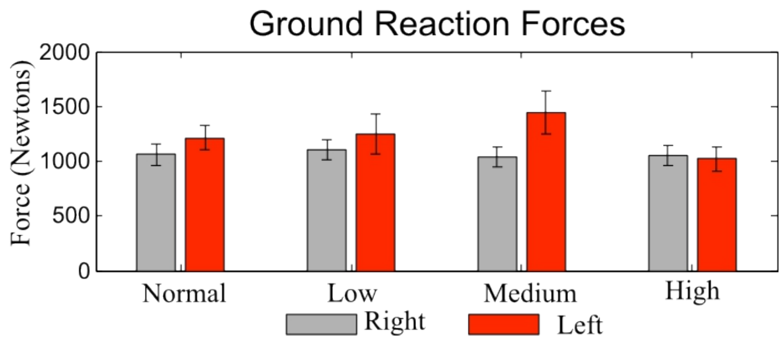


Figure 4.8: Ground Reaction Force Graph. For Every Trial on the CAREN System

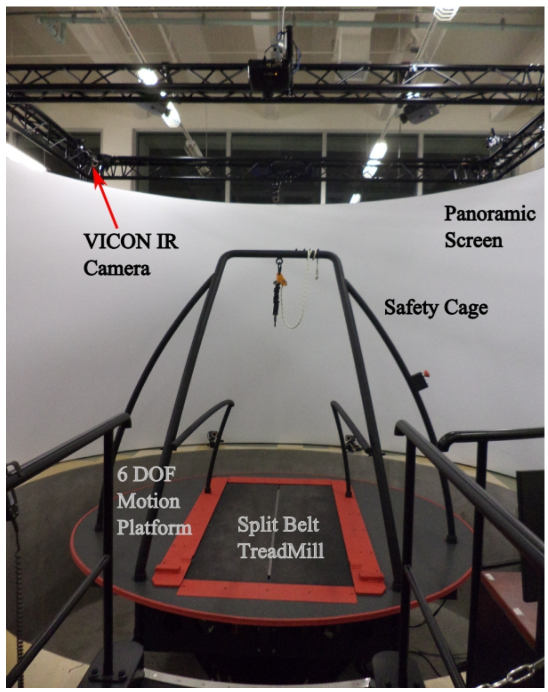


Figure 4.9: CAREN System

The asymmetric PDW model in this case can simulate variation in knee location, thigh width, shank width, damping and stiffness of the knee, and leg lengths. This versatile computational PDW model is purely driven by dynamics and does not have a neural feedback component to it. The comparison of kinetic and kinematic data from the prosthetic simulator to the model will show the impact of neural feedback in compensatory motion during walking. This test between neural feedback and pure dynamics will help in the design of better prosthetic systems.

CHAPTER 5: FUTURE WORK AND CONCLUSIONS

This research project has shown that an asymmetric prosthetic simulator fitted on able-bodied subjects can bring about changes in gait. It also showed that the individuals wearing the prosthetic simulator were compensating in different ways. Different compensation patterns are typical for individuals with transfemoral amputation. This is the first prototype of this design, and future iterations have to be tested with a larger population to show statistical significance of the results reported. This research can be further expanded to look into compensatory kinematics and kinetics at the hip joints, knee, and ankle joint of the healthy leg. The CAREN system also poses the human body model which can be used to identify the muscle groups used to compensate for the prosthesis.

The simple knee mechanism and kinetic foot shape have the potential to reduce the cost of passive prosthetics. Future design will concentrate on weight reduction, comfort, modularity in components, and focus on reducing energy costs during walking. Figure 5.1 shows a modular design concept of both the prosthetic simulator and a prosthetic leg with a socket. The designs are also fitted with symmetric locks that can be opened when the person needs to sit. Modularity allows the user to have multiple positions of the knee location. Modularity also allows for a lot of room for customizing the prosthesis. Another interesting use of the prosthetic simulator can be a form of hands free crutches with knee joints. This will allow the crutch user to have a more natural gait. A more in-depth investigation into the kinetic foot shapes is required to identify a foot shape that will work for most people. An asymmetric transfemoral prosthesis that can be fitted on amputees is also in the pipeline.

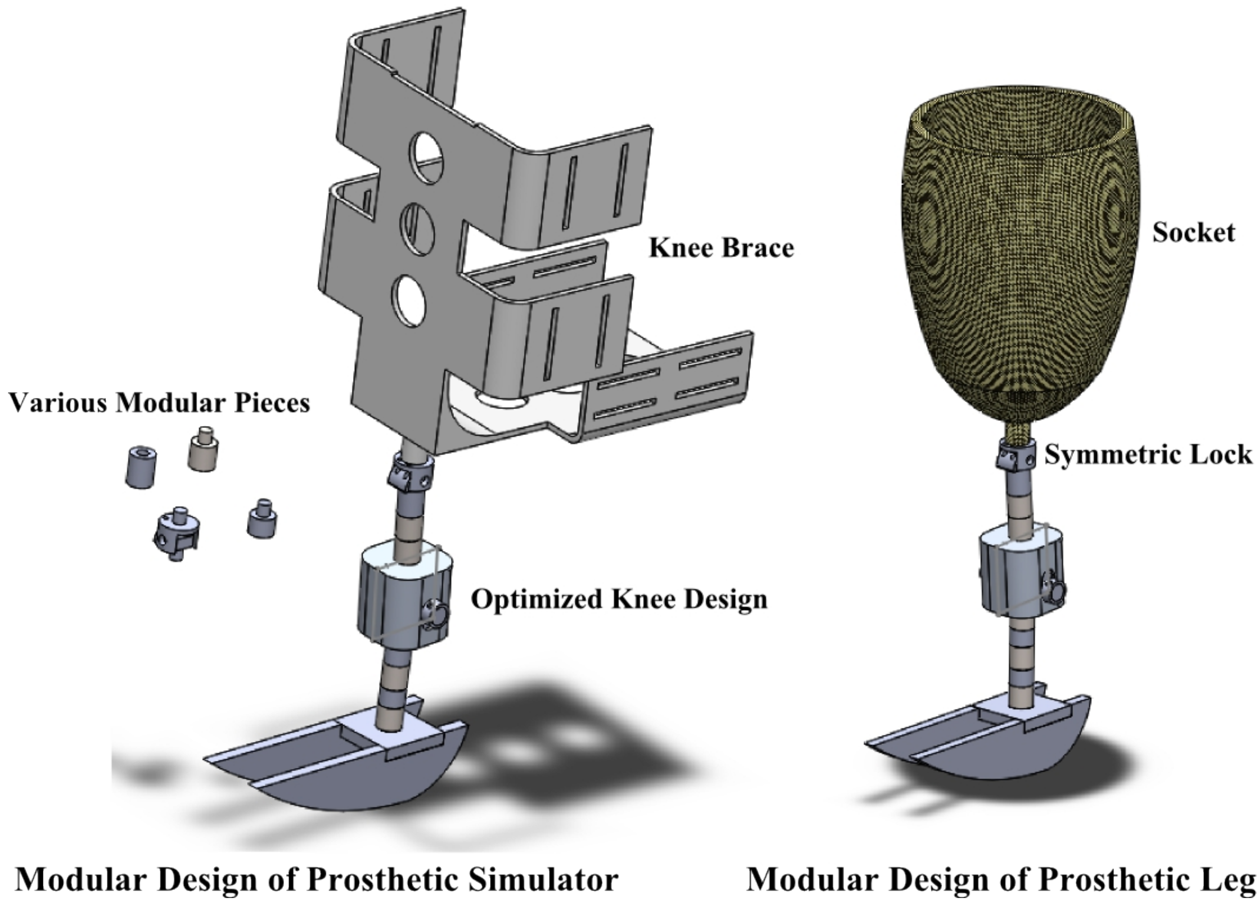


Figure 5.1: Future Modular Design Concepts

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APPENDICES

Appendix A: Pictures

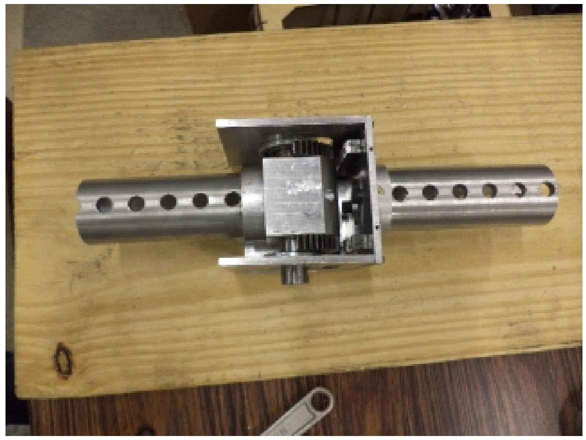

Figure A.1: Knee Assembly Without Collars

Figure A.2: Knee Assembly Dismantled

Appendix A: (continued)

Figure A.3: Knee Assembly Dismantled Isometric View

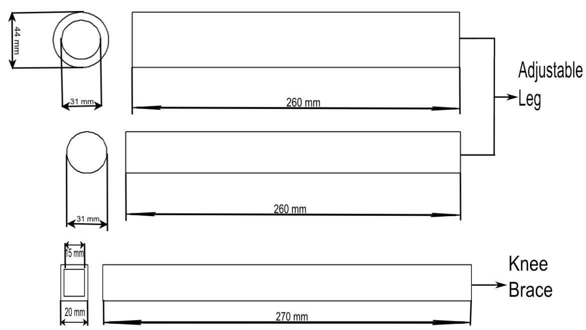

Figure A.4: Aluminum Raw Material 2D Drawing

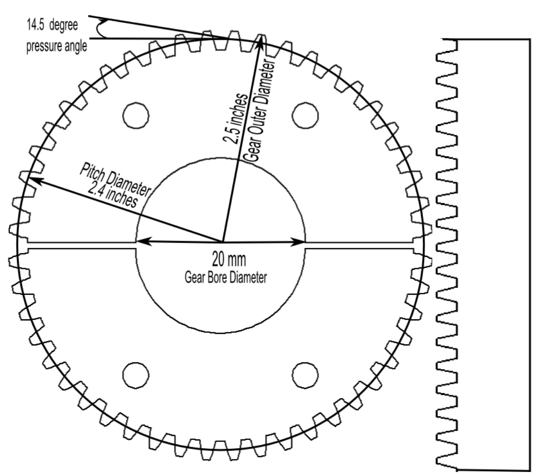

Figure A.5: Spur Gear and Gear Rack 2D Drawing

Appendix B: Graphs for All Trials

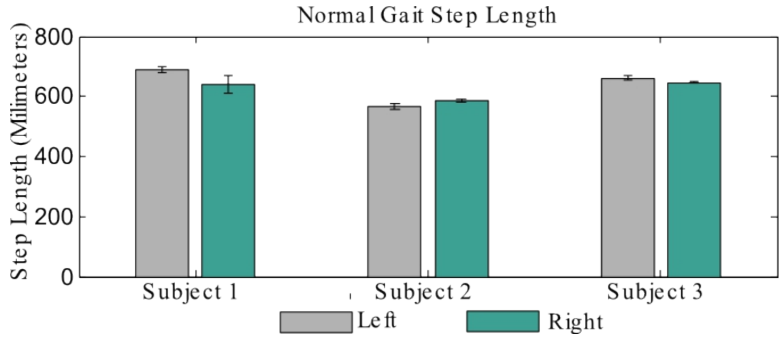

Figure B.1: Normal Walking Step Length Graph for All Subjects

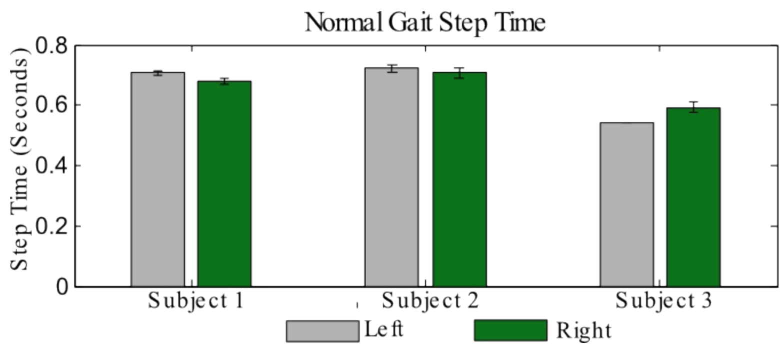

Figure B.2: Normal Walking Swing Time Graph for All Subjects

Appendix B: (continued)

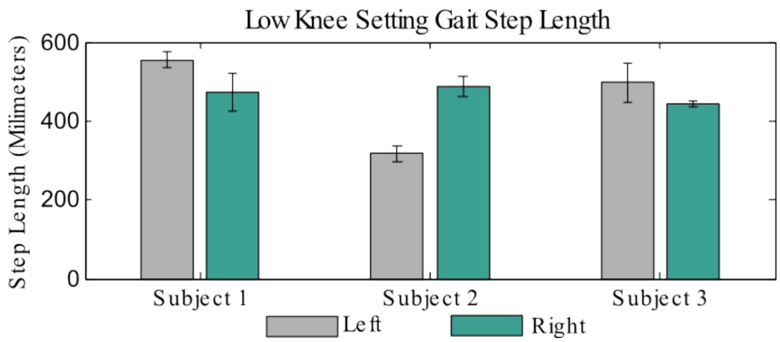

Figure B.3: Walking with Knee at Low Setting - Step Length Graph for All Subjects

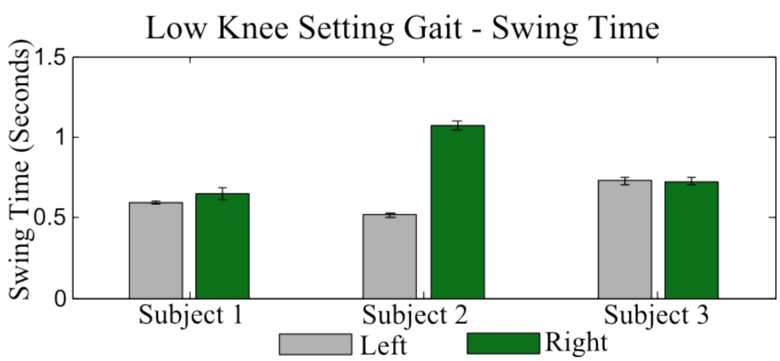

Figure B.4: Walking with Knee at Low Setting - Swing Time for All Subjects

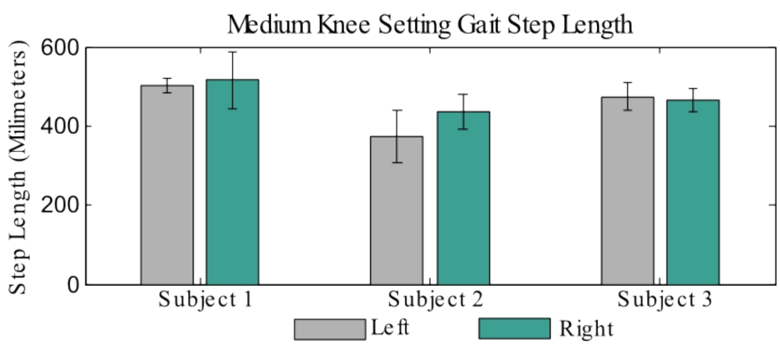

Figure B.5: Walking with Knee at Medium Setting - Step Length Graph for All Subjects

Appendix B: (continued)

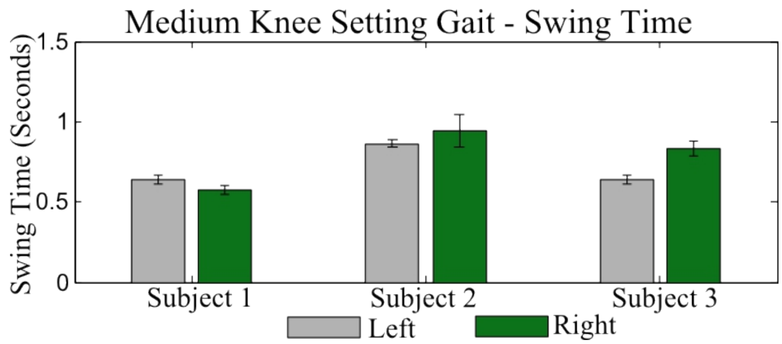

Figure B.6: Walking with Knee at Medium Setting - Swing Time for All Subjects

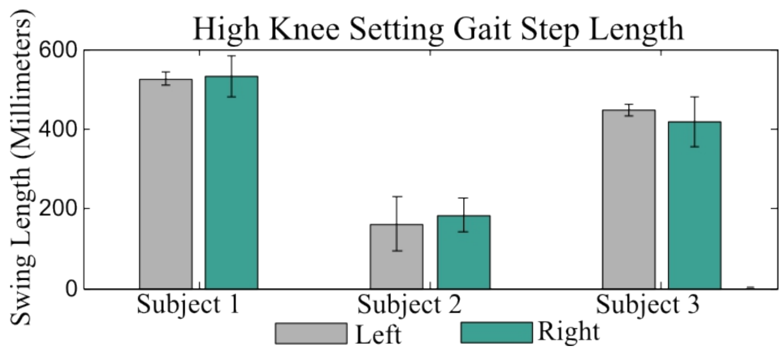

Figure B.7: Walking with Knee at High Setting - Step Length Graph for All Subjects

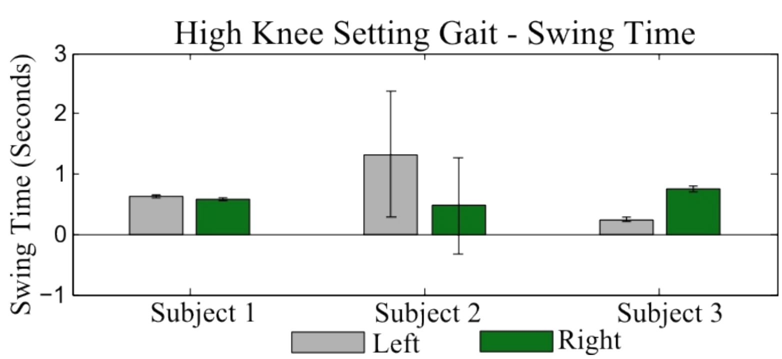

Figure B.8: Walking with Knee at High Setting - Swing Time for All Subjects