A PROSTHETIC DEVICE FOR TRANSTIBIAL AMPUTATION

A Final Year Project Report

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ABSTRACT

Naturally, the human ankle is capable of providing enough net positive work during walking, and therefore enables a normal human to propel forward. For a trans-tibial amputee, unlike the natural arrangement, the passive ankle-foot prosthesis is not capable of any such thing, and therefore it cannot aid or restore the natural gait in any way.

To enable the amputee to walk again, in a natural gait, a prosthesis capable of doing net positive work is required. The need for such powered prosthesis has been recognized due to the research on this matter but only one group of scientists so far has successfully developed and maintained that such a device which enables an amputee to have a natural walking gait. The main challenge in developing such prosthesis is to make it comparable to the actual human weight and size, and still be able to provide enough torque and power in order to follow the natural gait for a normal human.

Our work presents the design of mechanical components and their control for the working of a motorized ankle-foot prosthetic device which is capable of doing net positive work thus enabling the amputee to follow a natural walking gait. The basic model has a spring connected in parallel with a Series Elastic Actuator. The spring used is a unidirectional spring. The entire assembly approximates the normal human ankle size and weight and can provide enough torque and power in order to mimic normal human walking behavior. A control scheme is implemented to enable the prosthetic to follow the normal human gait.

ACKNOWLEDGMENTS

This work is a result of combined efforts and contributions of many people, who valued the work with their time and effort.

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Our parents have been a source of constant support and motivation throughout the degree. None of this would have ever been possible without their tireless efforts.

Most importantly we are thankful to the Almighty Allah that He blessed us with this opportunity and gave us strength to overcome any hardships that came our way.

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ABBREVIATIONS

ESR	Energy Storing and Returning
CD	Controlled Dorsiflexion
CP	Controlled Plantarflexion
DD	Direct Drive
LiPo	Lithium Polymer Batteries
PEA	Parallel Elastic Actuator
PP	Powered Plantarflexion
SEA	Series Elastic Actuator
SP	Swing Phase

NOMENCLATURE

K_{ankle}	Ankle stiffness
K_{CD}	Spring constant during CD phase
K_{CP}	Spring constant during CP phase
K_p	Parallel Spring constant
r_p	Moment arm
R_{total}	Total transmission ratio of the system
Text:	Ankle angle torque
T_h	Normal human ankle torque
T_{pp}	Torque during PP phase

CHAPTER 1: INTRODUCTION

The below-knee prosthesis available commercially in the market today are passive, and therefore cannot provide any net positive output work. Also, the mechanical properties of such prosthesis remain fixed with changing terrains. The research on this subject indicates that the trans-tibial amputees using these have a lot of difficulties, which include lower walking speeds, higher energy requirement, higher metabolic rates, and abnormal gait patterns compared to healthy individuals. For example, trans-tibial amputees roughly spend around 30 percent more energy to walk at the same pace as normal healthy individuals.

There are a lot of differences between a natural human ankle and the conventional ankle prosthesis. Most important of all, a normal human ankle performs more positive work in its operation, which is due to the energy provided by the muscles, particularly at normal to fast walking speeds. Researchers have suggested that the passive prostheses not being able to provide positive output work is the main reason of such difficulties for below knee amputees.

This is the main motivation to develop a powered ankle prosthetic device which can provide the amputee with the required net positive work during walking. We believe that such a device can have a significant impact on improving trans-tibial amputee ambulation, like normal walking speed, symmetry and normal walking metabolism.

With a market of over $5 billion, the prosthetic limbs market is growing faster than ever. Moreover, 1 million people lose a limb around the world every year. This has also motivated us to pursue this project for the betterment of mankind.

Problem Statement

Design and development of a cost-efficient artificial lower limb that mimics the working of a biological human ankle that can rehabilitate an amputee.

Objectives

Following are the objectives for our project:

- Design of a low cost below-knee device that can adapt human gait.
- System should be designed according to height and weight of the amputee to make him/her follow a normal gait.
- Approximates the biological ankle in its operation and provides jerk resistance.

CHAPTER 2: LITERATURE REVIEW

Innovation of a powered ankle has been discussed for more than 20 years. However, only a single attempt to develop such a prosthetic device is mentionable. Even after the mechanism being built no further work has been done on improving the gait of an amputee. Recently, the development of quasi-passive ankle-foot prostheses has been the focus of a lot of research work and studies. Several attempts were made to improve the amputee gait enabling the amputee to walk like a normal person.

Finally, researchers at MIT, including Hugh Herr, worked on a practical model that mimics the normal human gait, and also provides enough positive work to be used an amputee; which is the main inspiration behind our project. Given below is a summary of our study.

Basic Types of Prosthesis

Passive Prosthesis

Passive prosthesis have no kind of machinery either mechanical or electrical. Also, moving parts are absent in such prosthesis. It only serves the aesthetic purposes, and does not fulfill or provide any aide in restoring the daily routine tasks.

Figure 1: Passive Prosthesis

ESR (Energy Storing & Returning) Prosthesis

ESR usually consist of an energy storing and restoring element such as a mechanical spring. Energy Storing & Returning (ESR) take the form of a carbon fiber leaf spring with a deflecting keel component. Commonly used (ESR) prosthetic feet are unable to help maintain a natural human gait as they cannot supply torque or power.

Basically, it only has a limited amount of working efficiency and feasibility; and is limited by the spring used. It cannot contribute any positive work to the system.

Figure 2*: ESR Prosthesis*

Powered Prosthesis

These are the type of prostheses that, unlike the passive and ESR ones, are actually able to do positive work on the body. There are broadly two methods to do that: 1) Electric powered, and, 2) Body powered.

Electric prostheses are those which are controlled by electrical signals. These signals are generated from the muscles. We can say that the existing muscles are used to control the prosthetic. A sensor is placed within the device. It can:

- Obtain electrical signals
- Translate the obtained signals into movements.

- Execute required operations.

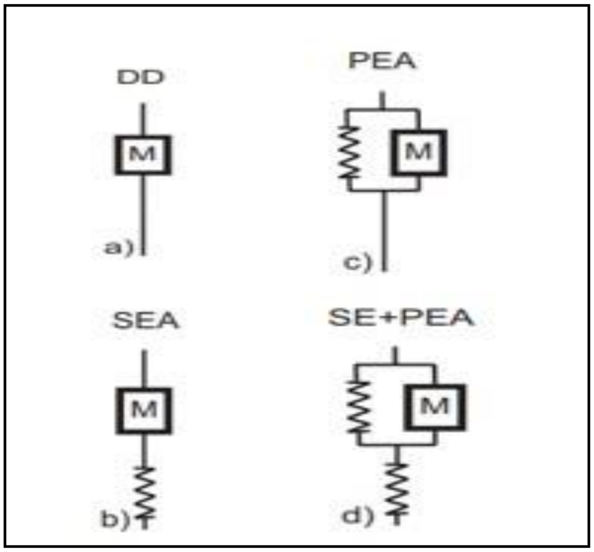

Figure 3: Electric Powered Protheses Models [8]

Table 1: Electric-Powered Prostheses Models

Type	Characteristics
Direct Drive (DD)	The basic powered model
Series Elastic Actuator (SEA)	Reduces Peak power Doesn’t Increase Energy Requirement
Parallel Elastic Actuator (PEA)	Reduces Peak Power Increases Energy Requirement
Series and parallel Elastic actuator (SE+PEA)	Reduces Peak power the most Increases Energy Requirement and complexity

Body-Powered Prosthesis

A *body-powered prosthesis* is different from the electrical one in a sense that it uses parts of body to control strings and cables which are used to control the prosthetic.

Figure 4: Body Powered Prosthesis

Hydraulic and pneumatic are the two methods that are commonly used for power transmission. Hydraulics utilize the flow of fluids under pressure to provide power. They can be useful in some applications, which cannot be dealt by people directly, or with automated systems.
Hydraulics work on Pascal's principle. Small force applied on a small area translates to a large force on a large area to keep the pressure constant.

Hugh Herr from MIT labs has published the pioneering work in the field of powered prosthetics. In our work, we will try to follow the models proposed by him, and physically implement these to develop a powered prosthetic that will be able to mimic the working of a normal human ankle.

Our prosthetic will be electrically powered ankle that will work using Series Elastic Actuator and springs and will work using finite state control scheme.

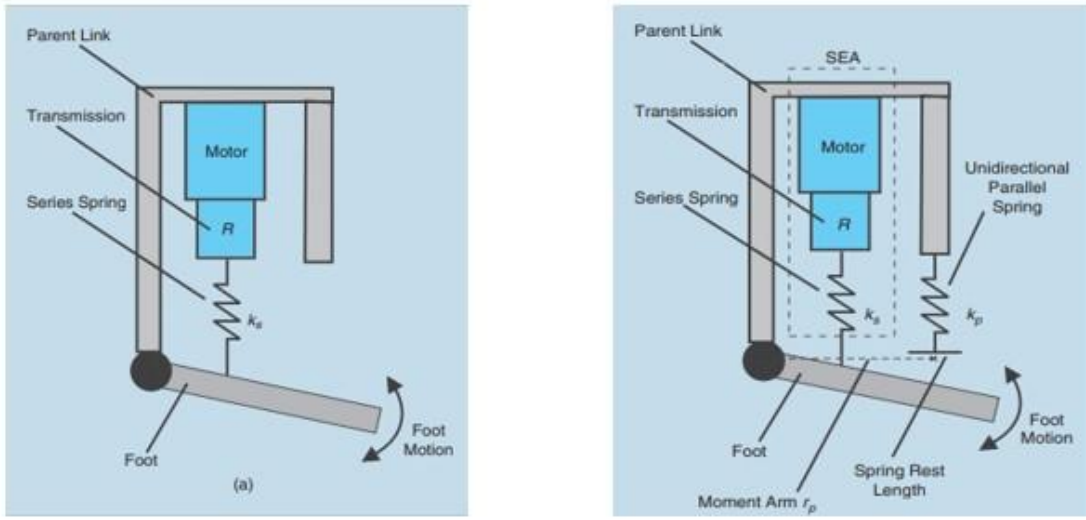

Figure 5: Schematic for powered ankle [1]

Approach

- Study walking behavior of a Normal Human Ankle-Foot
- Set target for a desired ankle movement.
- Model a mechanical system that mimics normal ankle behavior.
- Design the mechanical system to achieve required gait
- Design a control system to control the mechanical system to mimic normal human ankle behavior.

Engineering Challenges

For development of a powered prosthetic, there are two main hurdles. First problem is that it is not easy to build a prosthetic that matches the size and weight of the original limb.

Second, there is no target or reference against which the control of the prosthetic can be gauged to determine its effectiveness.

Outline

Our focus is to design a prosthetic using mechanical components that can mimic the behavior of human ankle, and a control system which can be used for its control.

CHAPTER 3: METHODOLOGY

The main purpose of our project is to make an amputee follow a normal human gait. The human ankle torque vs Ankle angle graph is given below in figure 6.

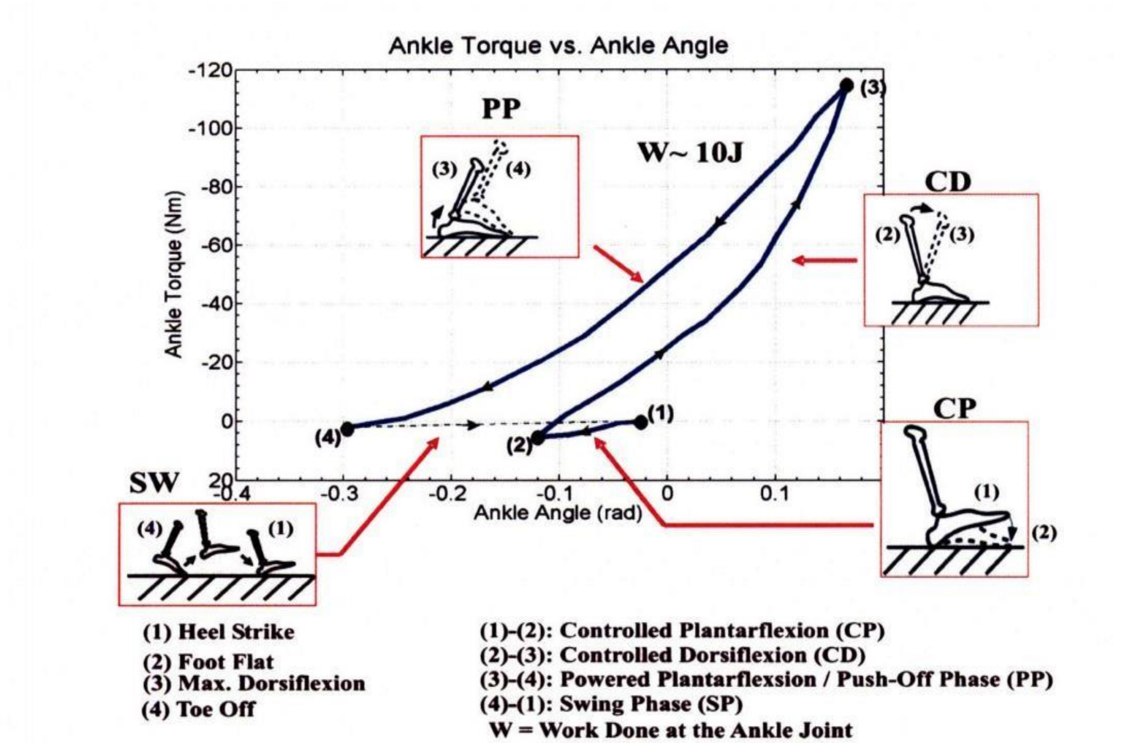

Figure 6: Normal human ankle angle vs Torque Curve [2]

As shown here, normal ankle behavior is given and it is divided into 4 steps, given above. An easy way to mimic the behavior of human ankle in the prosthetic is simply to mimic this quasi-static stiffness. Mechanically speaking, there are two main components in this.

1. A varying stiffness spring
2. A torque source

The stiffness of the spring changes in the same way as the human ankle in phases CP and CD.

The source of a torque acts in between points 4 and 3.

This model is further simplified, shown in the figure.

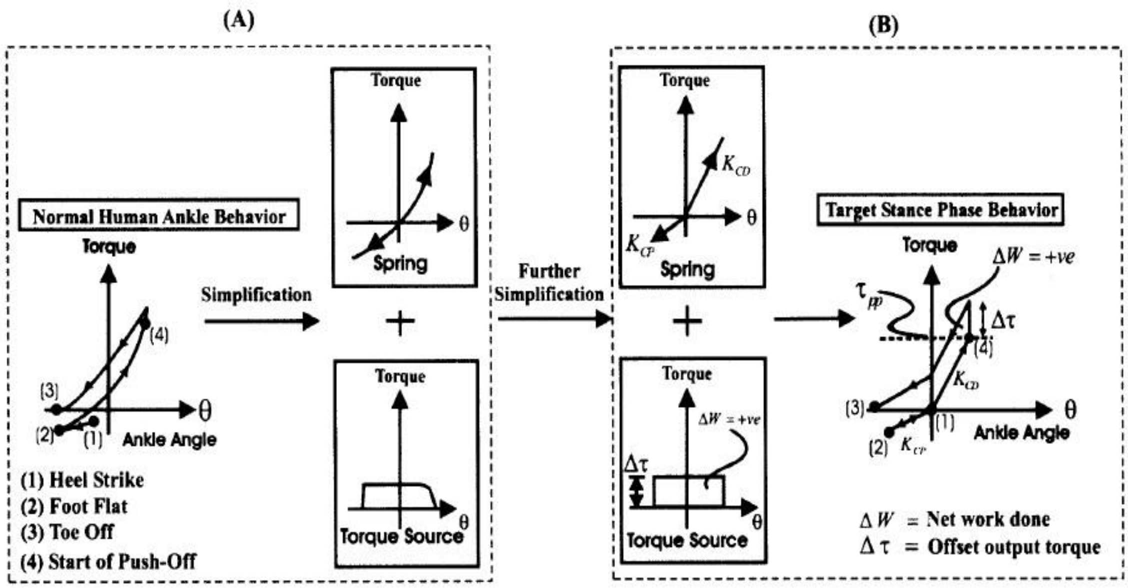

Figure 7: Normalized ankle angle vs Torque Curve [3]

As can be seen, the stiffness changes with the angle sign made by the ankle, as:

$$K_{ankle} = \begin{cases} \mathrm{K}_{CP} & \theta \leq 0 \\ \mathrm{K}_{CD} & \theta > 0 \end{cases}$$

During Push off phase, a constant offset torque T is provided. This is in addition to the K_{CD} which is provided during push-off.

This is the work which is done at the ankle joint. It is given as:

$$\Delta W = \Delta\tau\left(\frac{\tau_{pp}}{K_{CD}} + \frac{\Delta\tau}{K_{CP}}\right)$$

T_{pp}/K_{CD} indicates the angle at which torque is applied while T/K_{CP} represents the point at which application of torque is stopped.

Design Goals of Prosthesis

It should have a weight and height comparable to the normal human ankle. It must be able to provide required torque during push-off. It must change is stiffness according to the normal human behavior given in the graph above.

The system should be able to control the joint position, setting the ankle to its original position after the swing phase is over. The mechanism should be able to stand the body weight, and provide tolerance to shocks.

Size and Weight

[4]As a rough estimate, we take the weight of the missing limb to be about 2.5-3.0 Kg and height from 20-30 cm.

Range of Joint Motion

[5] gives us a max Plantarflexion of about 20-25 deg and max Dorsiflexion of about 10-15 deg.

Torque and Speed

[6] [2]Peak velocity 5.2 rad/s peak torque 140 Nm peak power 350 W. Prosthetic is designed such that it brackets these peak Torque Velocity and Power requirements.

Torque Bandwidth

It gives us how fast we have to output particular values of torque in order to meet our requirements.

[2] For human ankle, a torque value between 50 – 140 Nm, frequency of 3.5 Hz is required.

Our design goal is to bracket this value.

Net positive work

[6] [2] For a 75 kg person, at medium walking speed, value of net positive work done at ankle joint is 10 J

Offset Stiffness

It is the stiffness of the ankle joint during normal walking. It is important if we need to model the normal human walking behavior. It is found by taking the slope of the normalized human ankle angle vs torque curve, as given in [6] [2]

The value comes out to be 550 Nm/rad. It is applicable to medium walking speeds, i.e. between 1-1.8 m/s

The design parameters obtained by comparison with parameters for actual human walking are summed up as [2]:

Table 2: Design Specifications

Weight	2.5kg
Dorsiflexion	15 deg
Plantarflexion	25 deg
Maximum Torque	140 Nm
Maximum Velocity	5 rad/s
Maximum Power	350 watt
Torque Bandwidth	3.5 Hz
Total Work Done	10J at 1.3 m/sec
Offset stiffness during CD	550 Nm/rad

Mechanical Design

As it has been discussed earlier, our system includes a spring which is connected in series with a motor and a transmission. This assembly is called a series elastic actuator. A parallel spring is used to share some of the work load. A foot component is attached at the end. Conceptual design has been shown above.

Thus, we have our mechanical components:

- DC motor
- Transmission
- Series Spring and Parallel Spring
- Prosthetic Foot

Design Analysis

The Free body diagram for the model is given below.

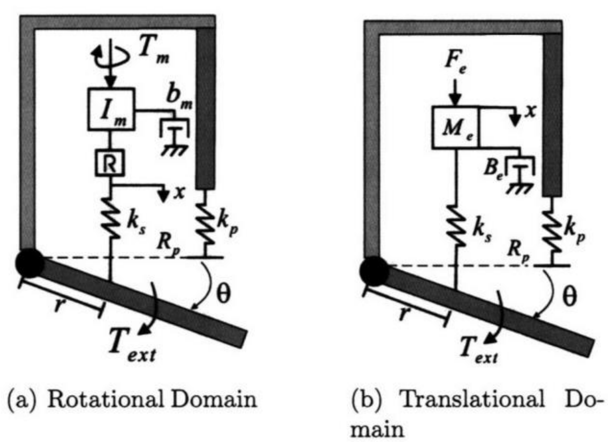

Figure 8: Free body diagram of model

The equation of motion is given a

$$
\begin{aligned}
M_e\ddot{x} + B_e\dot{x} &= F_e + F_s \\
F_s &= k_s(r\theta - x)
\end{aligned}
$$

The total external torque or total joint torque is given as:

$$T_{ext} = \begin{cases} \mathrm{rF}_s & \theta < 0 \\ \mathrm{rF}_s + R_p k_p \theta & \theta \geq 0 \end{cases}$$

Steady State Analysis

Steady State analysis is used to give maximum torque/power and speed characteristics. We will study the effects of actuator saturation and Transmission characteristics t to maximize in order to match the behavior of the ankle. Thus, the effects of springs, dampers as well as frictional components is not included.

Using this assumption the ankle joint torque becomes $\mathbf{T_{ext} = rRT_m}$

The performance of a motor is bound by its maximum torque and speed characteristics.

Hence, we have:

$$T_m(\omega) \leq T_m^{max} - \omega(\frac{T_m^{max}}{\omega^{max}})$$

If R_{total} = rR, where R_{total} is the total transmission ratio of the system.

Then, we have:

$$T_{ext}(\dot{\theta}) \leq R_{total}T_m^{max} - R_{total}\dot{\theta}(\frac{T_m^{max}}{\omega^{max}})$$

We can see that our design goal is to always have T_{ext} which is provided by our motor transmission, greater than T_h which is the normal human ankle torque. This can be represented as:

$$\begin{aligned} T_h(\dot{\theta}) &< T_{ext}(\dot{\theta}) \qquad \forall\dot{\theta} \\ &\leq R_{total}T_m^{max} - R_{total}\dot{\theta}(\frac{T_m^{max}}{\omega^{max}}) \end{aligned}$$

This gives us our primary goal for the prosthetic. The selection of motor and transmission should always justify the above equation.

Dynamic Analysis

In this section, the output acceleration of the system and large force bandwidth are studied. Output Acceleration:

For output acceleration, we are concerned with how fast the prosthetic can output its joint torque. Here, we are concerned with maximizing the output acceleration of the system.

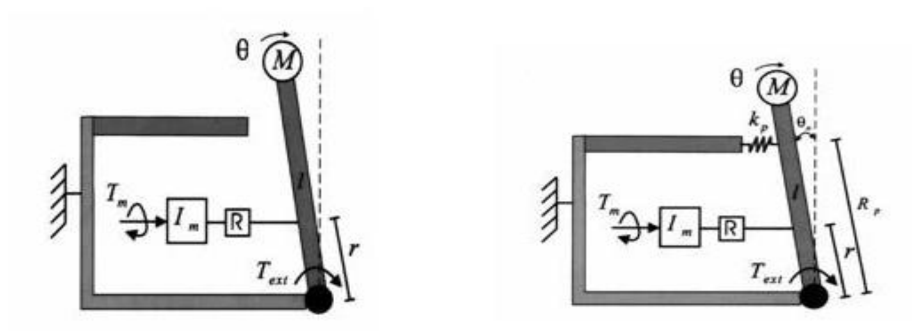

Figure 9: Free Body Diagram of powered ankle schematic [1]

The free body diagram is given in figure 9. The dynamic model of the system is given by:

$$\ddot{\theta} = \frac{T_m R_{total}}{Ml^2 + I_m R_{total}^2}$$

Where $\mathbf{R_{total} = rR}$

Differentiating the above equation wrt. R_{total} gives the optimal transmission ratio.

$$R_{opt} = \sqrt{\frac{Ml^2}{I_m}}$$

Max joint acceleration is given as

$$\ddot{\theta}_{max} = \frac{T_m}{2\sqrt{I_m M}}$$

Now, according to our model, if we incorporate the parallel spring here, the instantaneous acceleration is given as:

$$\ddot{\theta} = \frac{T_m R_{total} + K_p R_p \theta_o}{Ml^2 + I_m R_{total}^2}$$

It can be seen that addition of parallel spring increases the output acceleration for the same joint torque.

Large force bandwidth

While designing motors, one thing that we need to be careful about is actuator saturation. One method to take that into account is by measuring the large force bandwidth of the system. The frequency at which the system can oscillate at maximum motor force is known as large force bandwidth. Large force bandwidth is reduced significantly with series elasticity. Thus, the stiffness of the spring is directly proportional to the large force bandwidth. Hence, while designing the spring, this fact is kept in mind.

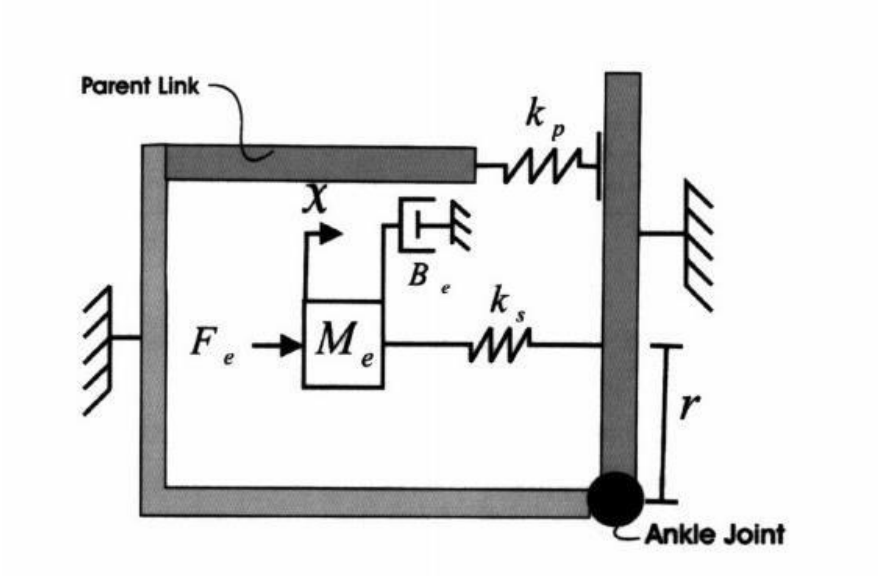

Figure 10: Free Body Diagram of powered ankle schematic [1]

The transfer function of the system in figure 10 is given as

$$G_{fixed}(s) = \frac{F_s}{F_e} = \frac{k_s}{M_e s^2 + B_e s + k_s}$$

$$\frac{F_s^{max}}{F_{sat}} = \frac{k_s}{M_e s^2 + (B_e + \frac{F_{sat}}{V_{sat}})s + k_s}$$

Thus, it can be seen that large force bandwidth depends on the choice of motor, transmission and springs.

Design Procedure

The designed procedure can be summed up as follows.

- Selection of motor and transmission to fulfill the steady state torque and velocity requirements

- Check whether motor and transmission selected can satisfy the output acceleration requirements.
- Select a parallel spring to share the load and increase instantaneous output system acceleration.
- Select a series spring to meet the large force bandwidth criteria.
- Design and analysis of mechanical components to support the normal human load during walking.

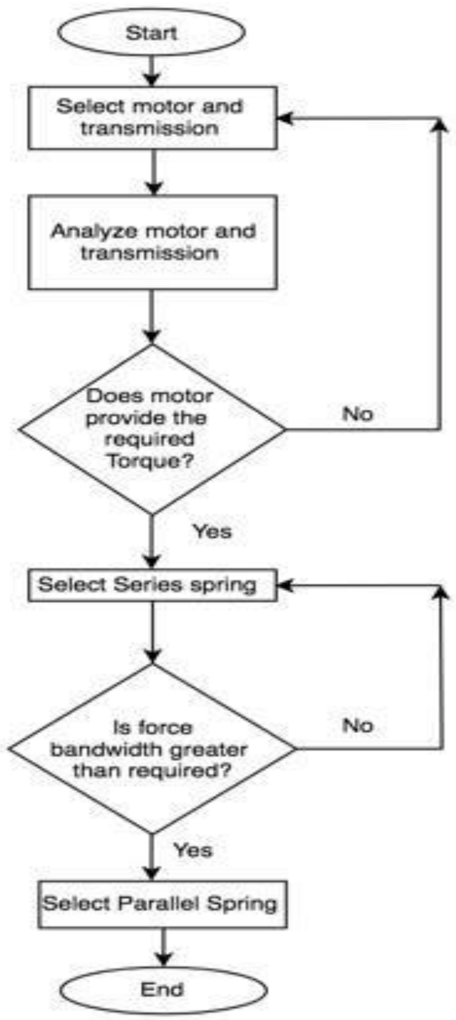

Figure 11: Flow chart for design procedure

Series spring selection

Series Spring is selected keeping in view the criteria discussed above. The large force bandwidth of the system should be at least 2-3 times greater than that required for the system. Keeping that in view, spring was selected with K_s = 1200 kN/m. This gives us a large force bandwidth 3 times greater than our requirement.

Parallel Spring Selection

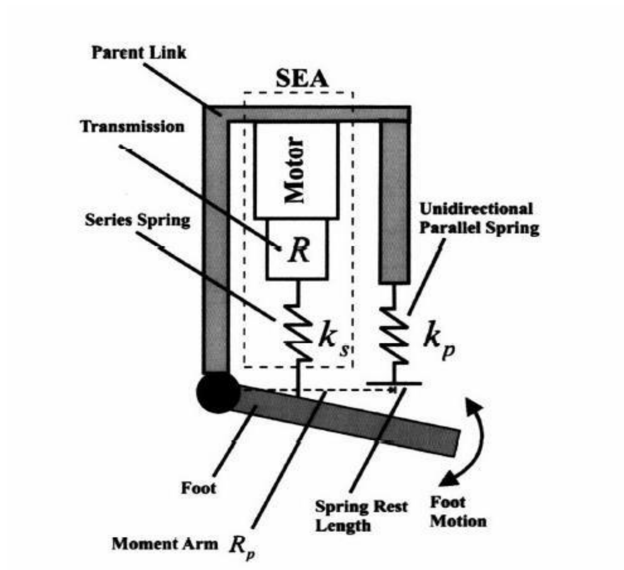

Figure 12: Free Body Diagram of powered ankle schematic [1]

Here,

$$K_p^\tau = (k_p)(R_p)^2$$

Thus, we need to select the moment arm and stiffness of the spring to get the offset stiffness given above in the design parameters. For our case, calculations were done and moment arm 0.022 m and spring constant is 770 kN/m. This gives us offset stiffness of 385 rad/sec. Our offset stiffness value given in design parameters was 550 rad/sec. Thus, Series Elastic Actuator used supplements the stiffness of parallel spring.

Motor Selection

Motor was selected based on our criteria for torque and velocity discussed above. Two motors were found usable for our requirements of torque and velocity. Maxon RE 40 and Mabuchi RZ 735

These were compared and Maxon RE 40 was selected after parametric analysis, details are given in the appendix.

Transmission selection

For transmission, Ball screw was used to convert rotational motion of the motor into linear motion. Worm gears can also be used for this purpose, however, ball screw was found to be better suited for our requirements.

Figure 13: Ball screw assembly

The formula for calculation is given as

$$Md = \frac{P_h}{2\pi} \cdot \frac{1}{\eta . r_t} . F_{feed}$$

The parameters are summed up and given in the table 3.

Table 3: Ball Screw Design Parameters

Transmission Ratio	133
Ball screw	Thomson 8103-448-023
Lead	3mm
Pitch	3mm
Static Load Rating	8.6 kN
Dynamic Load Rating	4.8 kN
Ball screw transmission R_t/r	3560
Belt Transmission r_t	1.7

Viability was checked according to the criteria given above, which was satisfied.

For the combination of motor and transmission, a graph was also plotted to check the viability of the system.

Data for normal human ankle was taken from literature.

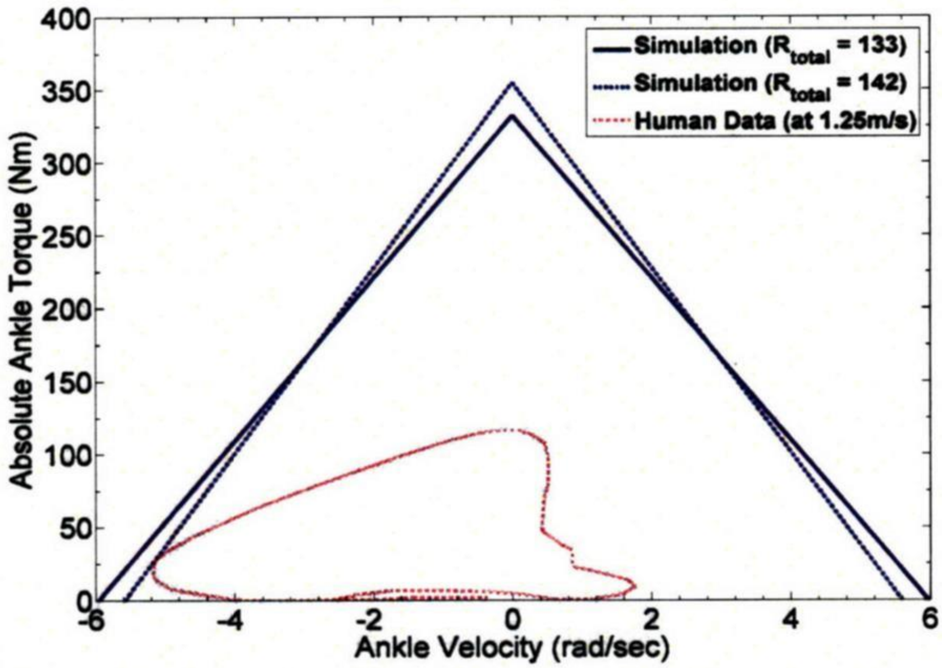

Figure 14: Torque vs velocity curve for selected motor [7]

Control System

Control of mechanical and electrical components, in order to mimic normal human ankle behavior comes under control system. This task is accomplished using a combination of controller with sensors, which provide data that allows the controller to act according to the situation. Finite state controller based approach is used.

Three main types of control strategies are used.

Torque Control

It is used to control when torque will be applied, according to the ankle angle and torque curve.

Position Control

This is used to return the ankle back to its original position in the swing phase of the ankle. Impedance Control

This is used to adjust the stiffness of the series spring, according to the requirements from the torque and ankle angle curve.

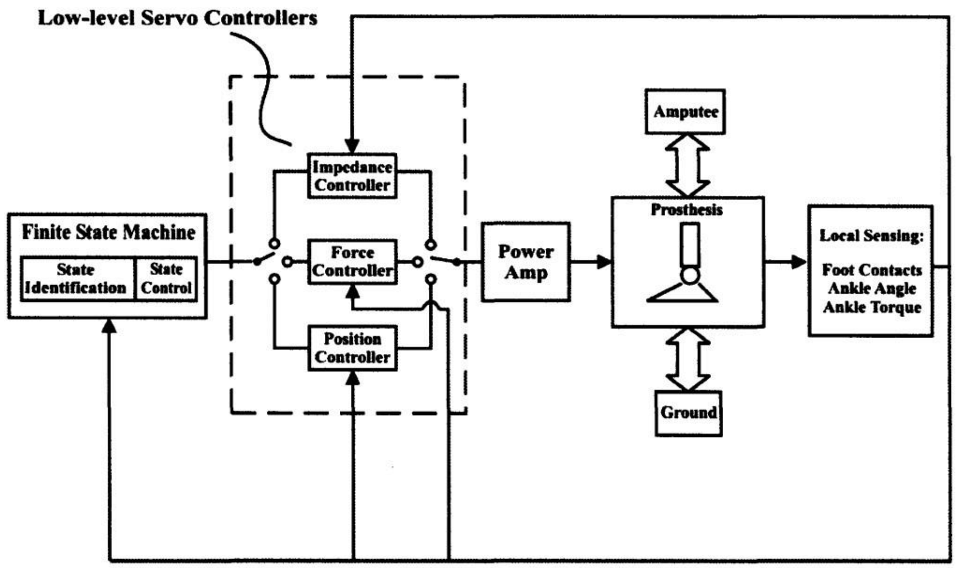

Figure 15: Schematic for control system [7]

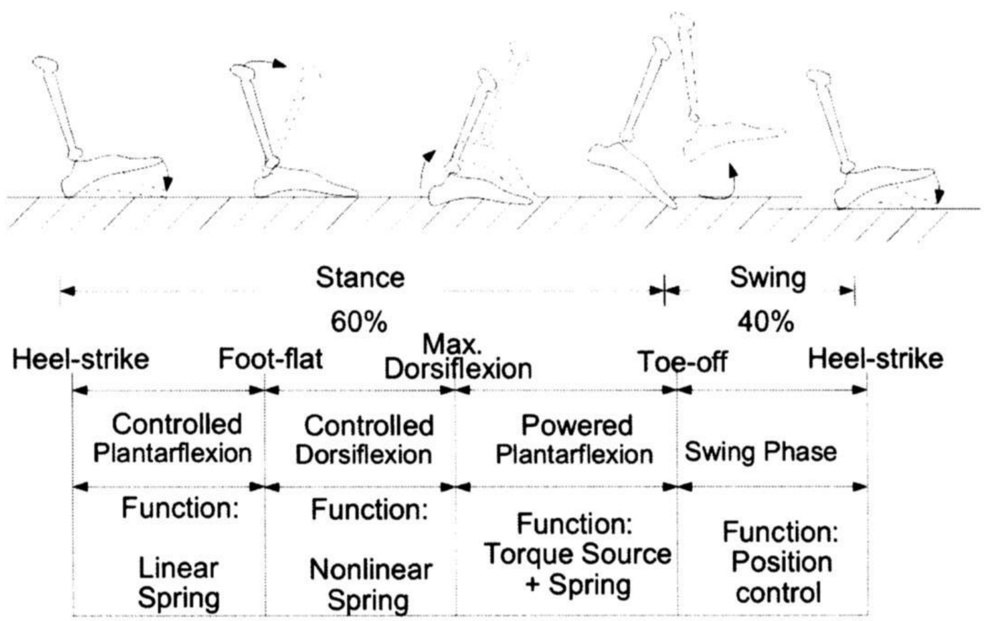

Figure 16: Stages of human walking [2]

Normal human ankle behavior is given in the figure above. It is divided into stance phase and swing phase.

- Stance Phase Control
- Swing Phase control

The behavior of the system, after applying our control scheme is given below.

Stance Phase Control

- CP begins at heel strike, ends at mid-stance, joint outputs a stiffness Kcp
- CD begins mid-stance and ends at PP or toe off.
- During CD the prosthesis outputs a joint stiffness K_{cd}
- During PP, push-off phase, offset torque is provided

Swing Phase Control

- Reset foot in initial position

Sensing for state transitions

- Heel Contact
- Toe Contact
- Ankle Angle

These parameters are used to sense the state transition. A combination of Pressure sensors and potentiometers will be used to detect heel contact, toe contact and ankle angle. The data from these will go into the controller for required action.

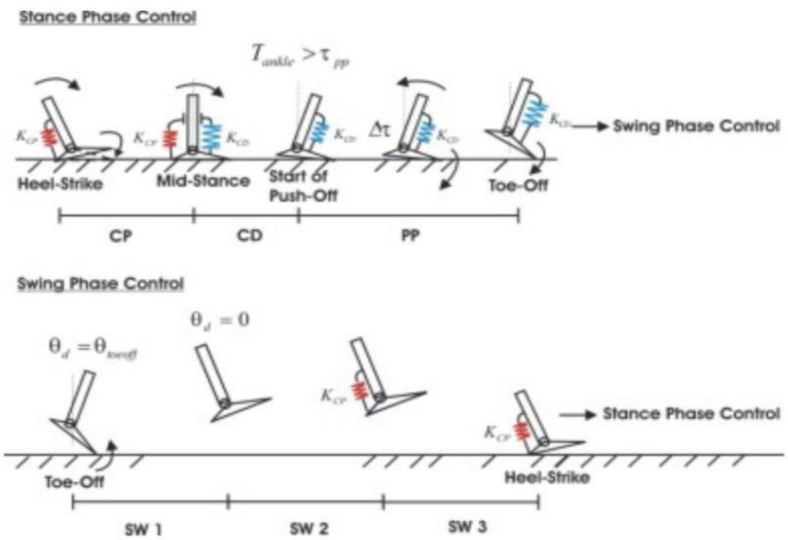

Figure 17: Phases of human walk [2]

This concludes the working of our control system, and this part of our analysis.

CHAPTER 4: RESULTS AND DISCUSSIONS

Mechanical component design:

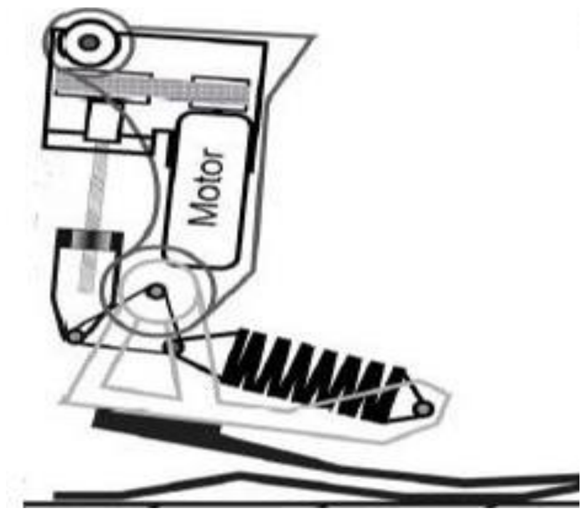

Figure 18: Powered ankle schematic

Figure 19: SolidWorks assembly for ankle prostheses

Dimensions were selected according to our requirement, and model was developed in Solidworks.

From the model, critical components were identified and stress analysis was performed to determine compatibility with normal use. The critical component here is the link, as the entire stress acts on it. To share the load, two links are used, so each has to bear only half stress. And links being critical components, are made of mild steel. The rest of the body (Upper housing, Bracket) is made of Aluminum to reduce the overall weight. Although no stresses act on the upper housing and the bracket under normal working conditions, as they share no load, their analysis is still performed, assuming the worst case scenario, where the entire weight acts on these parts.

Spring and ball screw were selected according to design criteria and rated load, so their analysis was not needed

Upper Housing:

Figure 20: CAD Model of Upper Housing

Stress analysis for static failure as well as buckling analysis was performed.

Buckling Analysis

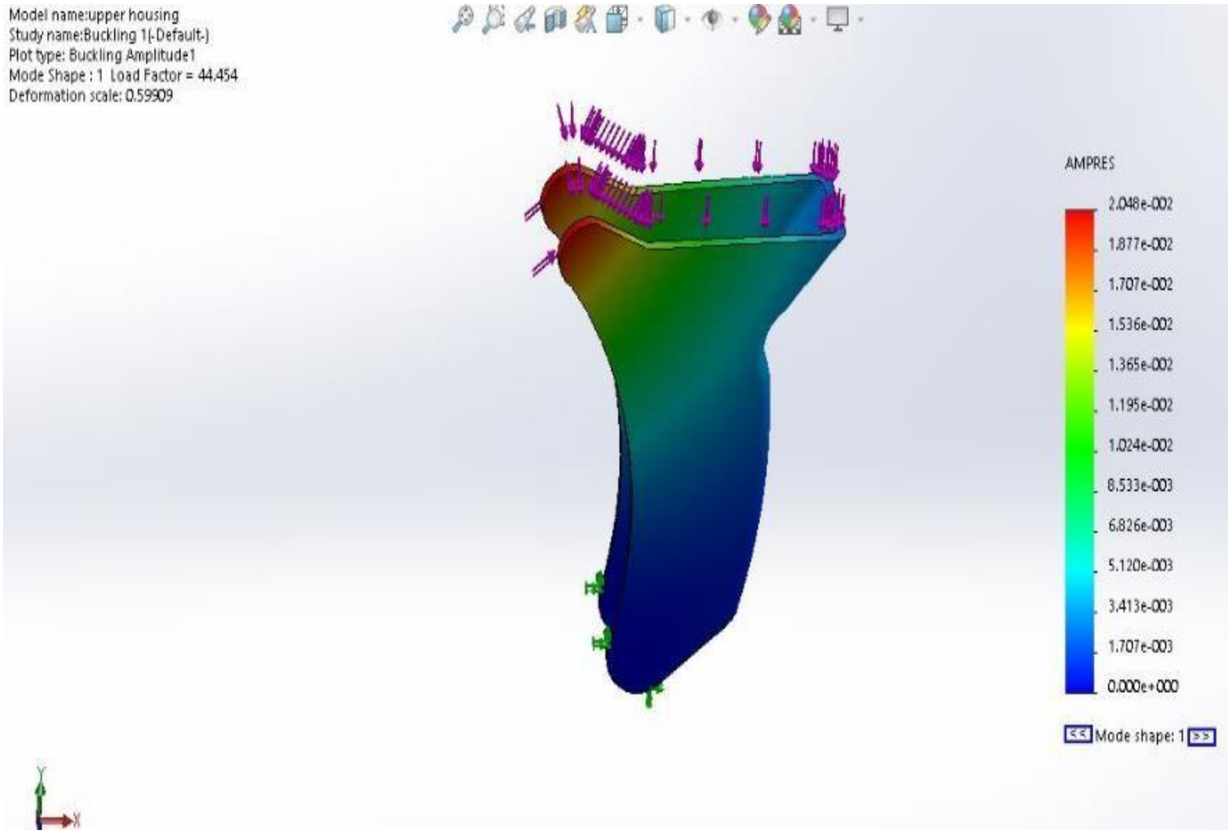

Figure 21: Buckling Analysis of Upper Housing

Here, a load factor of 44.45 suggests that the structure will never buckle at our loading conditions.

Static Analysis

Von Mises criterion was used to determine the conditions of failure and analysis was performed.

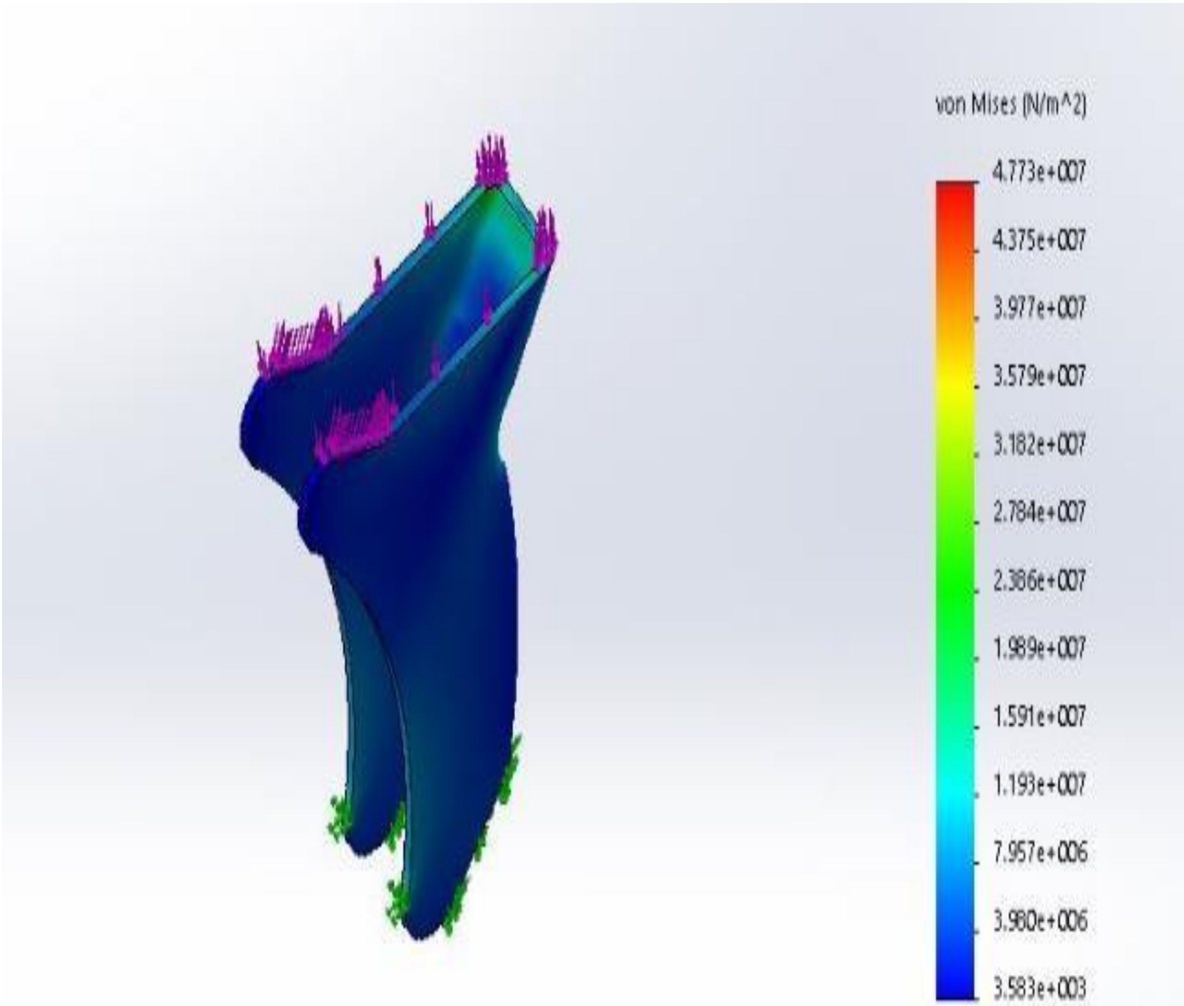

Figure 22: Static Analysis for upper housing

The material used is mild steel with a yield strength of 2.6 x 10^8 Pa. It can be seen the value shown here i.e. 4.7 x 10^7 Pa is well below the yield limit. Hence, material is safe.

Lower support

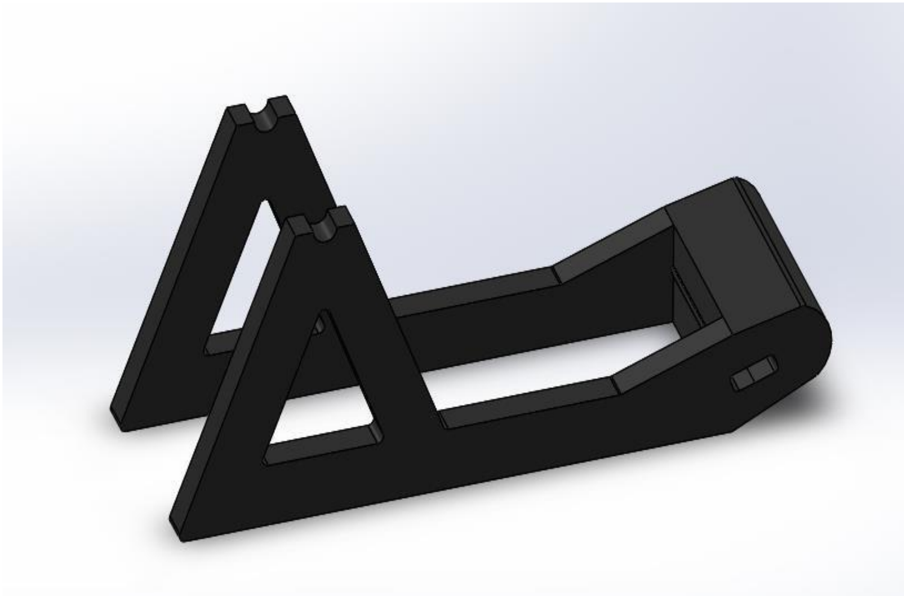

Figure 23: Lower Support CAD Model

Buckling

Stress Analysis for buckling as well as static failure was performed.

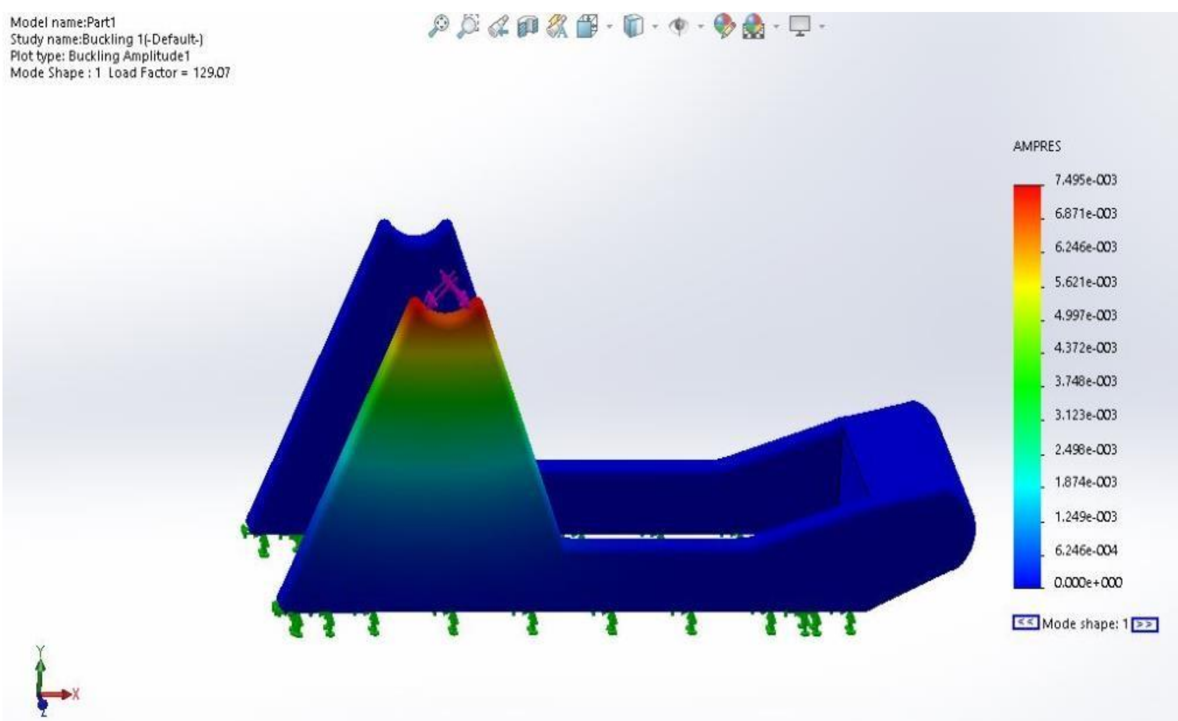

Figure 24: Buckling Analysis of Lower Support

In this case, for a total 75 kg person, half the load was applied on one face (both share equal loads). Here too, the high load factor shows that the material will not buckle at well above our required limits.

Static Failure

Von Mises criterion was used.

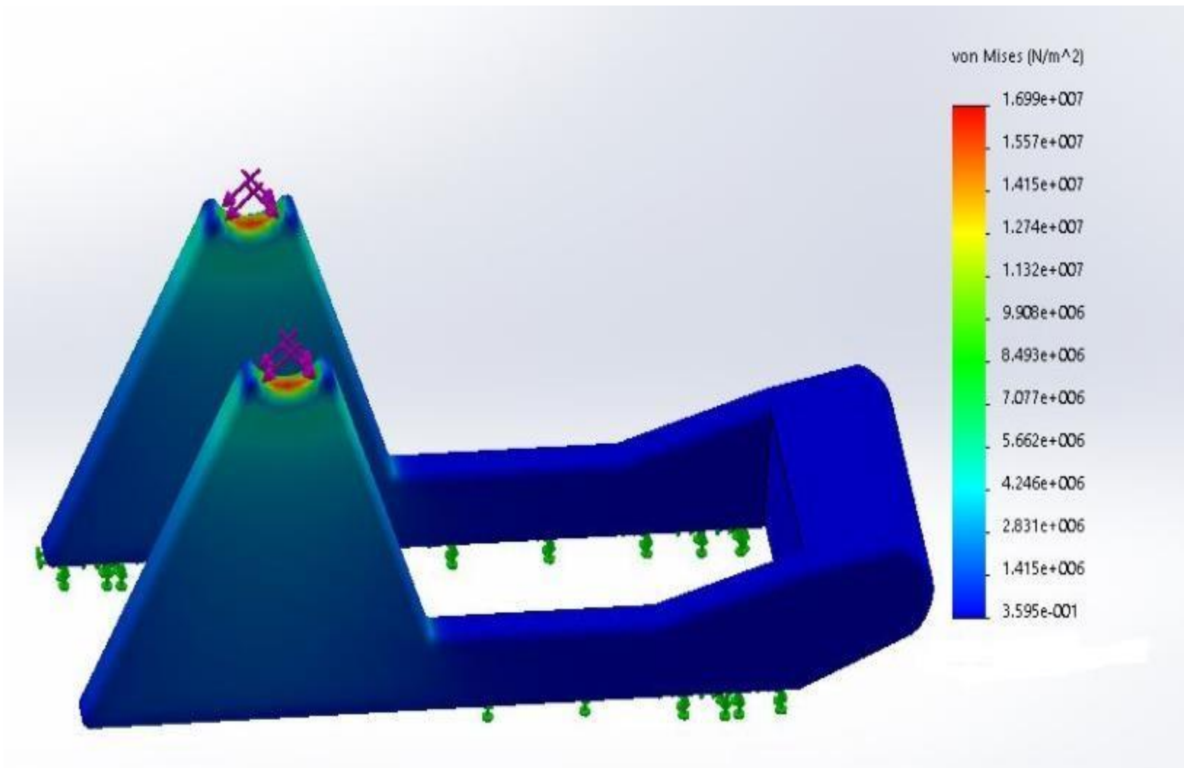

Figure 25: Static Analysis for Lower Support

The material used is Aluminum with a yield strength of 2.6 x 10^8 Pa. It can be seen the value shown here i.e. 1.6 x 10^7 Pa is well below the yield limit. Hence, material is safe.

Link:

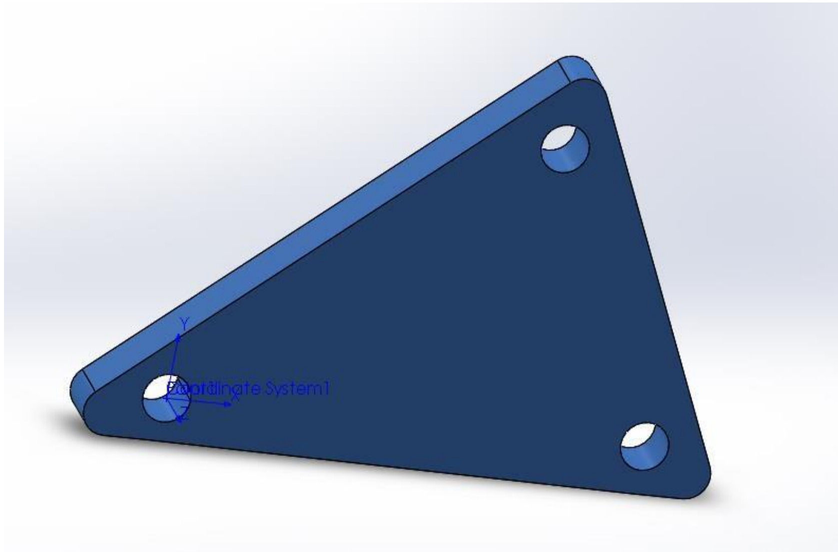

Figure 26: CAD Model of Link

Forces on each of these was calculated. Considering for the worst case scenario, one side was fixed and the force due to the entire motor torque was applied as the bearing load on the other side. Von Mises criterion was used for the static failure analysis was performed.

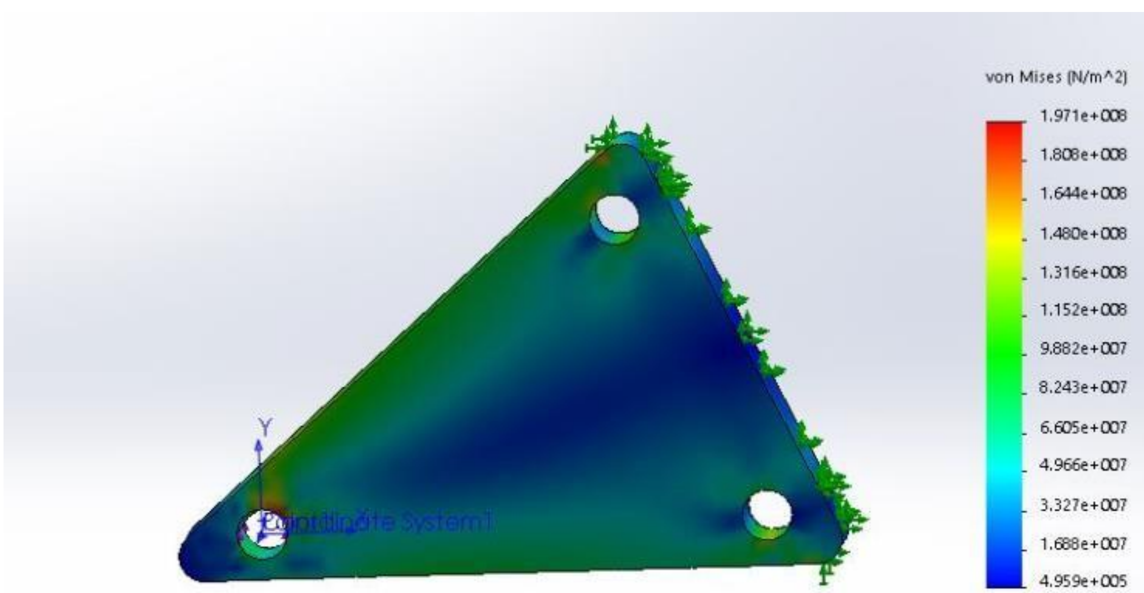

Figure 27: Static Analysis of Link

The material used is mild steel with a yield strength of 3.7×10^8 Pa. It can be seen the value shown here i.e. 1.9×10^8 Pa is below the yield limit. Hence, material is safe.

Also, the calculations performed in this case are for the worst case scenario. In actual world, these conditions will not exist.

Convergence Studies

Convergence studies were performed to ensure that the results presented here are converged. A simple method was used for each of these cases. Number of mesh elements were doubled for each case, and simulation was performed again. The difference for peak stress was found to be less than 2%, implying convergence.

From the above studies, we have all the required parameters. For the next phase, the product was manufactured and assembled, with a few changes based on the availability of the components and the market price. This will be discussed in the next section.

CHAPTER 5: CONCLUSION AND RECOMMENDATION

Our actual finished product is shown below in figure 28.

Figure 28: Completed physical assembly

Mechanical Design

The mechanical parts and links were manufactured according to our CAD model dimensions, which was verified by our analysis. However, a few changes were required. Obtaining the required motor and the ball spring proved to be difficult and expensive. Thus, instead of purchasing motor, ball screw and joining them, a pre-built linear actuator was purchased, which was rated for 1500 Nm load, and met our above mentioned design criteria. This saved us a lot of money as well as the trouble of finding and joining the required parts. This linear actuator was the connected to a spring to form the Series Elastic Actuator, which was used.

Power Source

LiPo batteries, rated 12 V and 2000 mAh were used as a power source. Arduino was also powered through a 6V battery, making the system completely mobile. However, for our testing purpose, Arduino was connected to laptop also to get the sensor data for observation. This is not needed for the actual operation of the ankle, and the system is completely mobile/portable.

Control

For control, the scheme proposed in our design section was used. Force sensitive resistors, placed at the toe and heel were used to detect contact. MPU 6050 (combination of accelerometer and gyroscope) was used to detect the angle at each point. These were then used to detect phase, and output torque according to the above referenced ankle angle and torque curve.

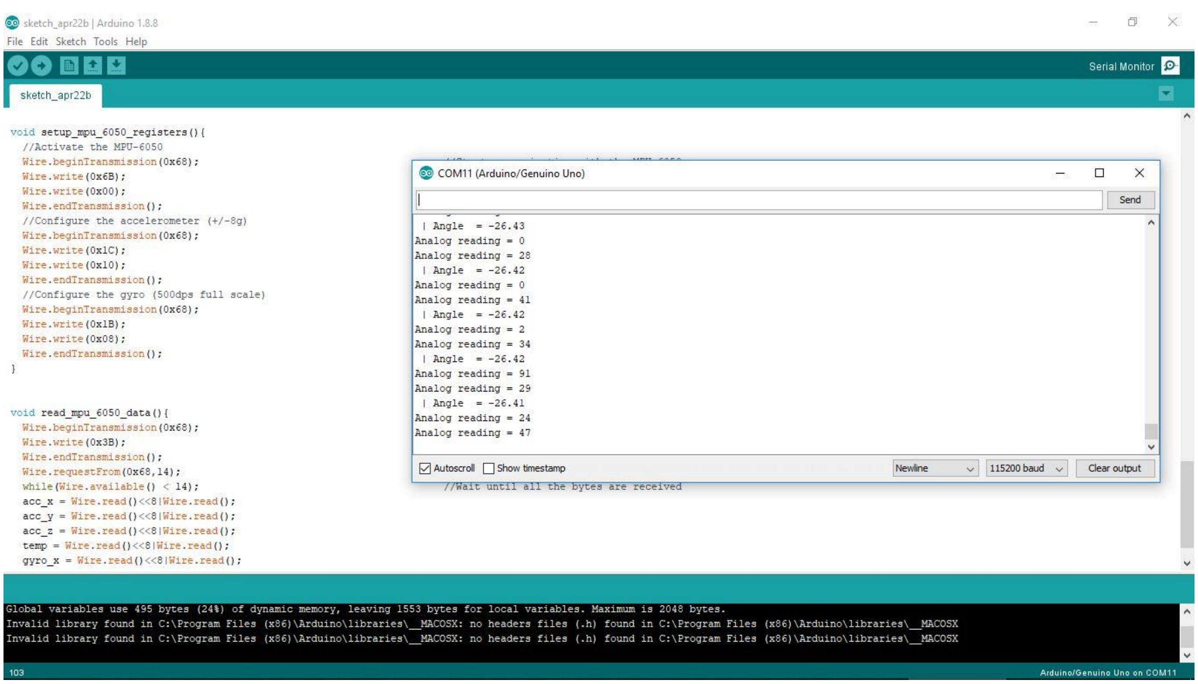

Figure 29: Pressure sensor and ankle angle monitor

Testing Rig

A testing apparatus was manufactured, and connected on top of the ankle, as shown in the figure. Normally, this would be placed under the amputated leg of the amputee and joined using special connectors. However, in our case, we did not have access to an amputee for our tests. So, the rig was modified in a way that it could be placed under the knee of a normal healthy person, to be used for testing.

Weight

The weight of the entire assembly, excluding the testing rig, is 2.5 kg, which fits our design criteria. Hence the ankle will not feel unnaturally heavy.

Testing

Tests were performed, with one of our group members wearing the ankle using the testing rig, and walking with it. The test was a success, and the prosthesis was successful in behaving like a normal ankle.

However, much more in depth results regarding the exact mimicry of the normal ankle angle and torque curves could not be derived. This is because the ankle was designed for a trans-tibial amputee. However, with our testing rig, by using this on a normal healthy person, the knee joint goes out of the equation. And hence, it becomes difficult for the testing person to control his balance, without the aid of the knee for positioning.

Price

The total price for development of our prosthesis was about 20,500 Rs. This is very cheap, considering the price of other powered, and even unpowered prosthetics in the market. Just as an example, the bionic ankle developed by Hugh Herr and MIT labs, costs about 40,000 $.

Once it leave the prototype phase and enters the product phase, the prices will reduces even further. A breakdown of costs is given below.

Table 4: Price of components

Component	Price (Rupees)
Linear Actuator	4200
Parts 3d Printing	8000
Aluminum Rod	450
MS Sheet	250
Nuts, Washers, Key	760
2 Drill Bits	140
Glue Gun + 5 Glue Sticks	700
MPU 6050	250
12 V Adaptor + 2 Pin Connector	410
12 V LiPo battery	1400
9V battery and connector pins	100
Al Plate	350
Force Sensitive Resistors x 2	1150
Machining of Parts	2000
Arduino Cable + Pins	310
Total	20,470

Recommendations and future work

Powered prosthetics, especially powered ankle prosthetics are a relatively new field. Research has been done in this field, especially by Hugh Herr of MIT labs. He is the pioneer of powered Ankle Prosthetics. Most of the design schemes proposed here are based on his work. However, this work is still in progress, and currently there is no powered prosthetic available in the market. Although the MIT Ankle developed by Hugh Herr in MIT labs mimics the normal human ankle behavior to an extent, the work is far from done. Also, it is extremely expensive, with the cost upwards of 40,000$. There is a lot more research work yet to be done to perfect it.

For now, our target is to manufacture an ankle that meets our criteria for suitable height, weight and power, and can allow rehabilitation of amputees, by closely mimicking the behavior of human ankle. We will implement an approach based on finite state controller.

Some of the aspects that can be considered in order to improve the powered ankle prosthetics are as follows:

Testing

The main thing that is left for future work in our project is the testing on actual amputees, and the collection of data from those tests to compare to the ideal human walking behavior. In our case, although we were able to develop a working prototype, all testing was done on healthy individuals, with a specially designed testing apparatus. One of the issues is that the prosthesis is designed for trans-tibial amputees, and their knees work. However, in our case, wearing the testing apparatus takes the knee out of the operation, and the wearer has problems with balancing and positioning. So the main task in the future will be the testing on actual amputees, and comparison of the results with amputees wearing other forms of prosthetics.

Total Energy Requirements

Further work can be done for optimization in order to minimize the total energy requirements of the system, in order to improve its efficiency and to provide a longer life per charge of batteries. As mentioned above, we cannot get accurate readings for the energy requirements unless we do the testing on actual amputees. Hence, this is also the main focus for future work in this project.

APPENDIX I: Comparative analysis for Parallel Spring:

Kl	Rp1	Rp2	Rp3	Rp4	rp5	rp6	rp7	rp8	Kr1	Kr2	Kr3	kr4	kr5	kr6	kr7	kr8
100000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	10	16.9	25.6	36.1	48.4	62.5	78.4	96.1
150000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	15	25.35	38.4	54.15	72.6	93.75	117.6	144.15
200000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	20	33.8	51.2	72.2	96.8	125	156.8	192.2
250000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	25	42.25	64	90.25	121	156.25	196	240.25
300000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	30	50.7	76.8	108.3	145.2	187.5	235.2	288.3
350000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	35	59.15	89.6	126.35	169.4	218.75	274.4	336.35
400000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	40	67.6	102.4	144.4	193.6	250	313.6	384.4
450000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	45	76.05	115.2	162.45	217.8	281.25	352.8	432.45
500000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	50	84.5	128	180.5	242	312.5	392	480.5
550000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	55	92.95	140.8	198.55	266.2	343.75	431.2	528.55
600000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	60	101.4	153.6	216.6	290.4	375	470.4	576.6
650000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	65	109.85	166.4	234.65	314.6	406.25	509.6	624.65
700000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	70	118.3	179.2	252.7	338.8	437.5	548.8	672.7
750000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	75	126.75	192	270.75	363	468.75	588	720.75
800000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	80	135.2	204.8	288.8	387.2	500	627.2	768.8
850000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	85	143.65	217.6	306.85	411.4	531.25	666.4	816.85
900000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	90	152.1	230.4	324.9	435.6	562.5	705.6	864.9
950000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	95	160.55	243.2	342.95	459.8	593.75	744.8	912.95
1000000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	100	169	256	361	484	625	784	961
1050000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	105	177.45	268.8	379.05	508.2	656.25	823.2	1009.05
1100000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	110	185.9	281.6	397.1	532.4	687.5	862.4	1057.1
1150000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	115	194.35	294.4	415.15	556.6	718.75	901.6	1105.15
1200000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	120	202.8	307.2	433.2	580.8	750	940.8	1153.2
1250000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	125	211.25	320	451.25	605	781.25	980	1201.25
1300000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	130	219.7	332.8	469.3	629.2	812.5	1019.2	1249.3
1350000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	135	228.15	345.6	487.35	653.4	843.75	1058.4	1297.35
1400000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	140	236.6	358.4	505.4	677.6	875	1097.6	1345.4
1450000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	145	245.05	371.2	523.45	701.8	906.25	1136.8	1393.45
1500000	0.01	0.013	0.016	0.019	0.022	0.025	0.028	0.031	150	253.5	384	541.5	726	937.5	1176	1441.5

The green region shows the spring constant values that can be used for our case.

APPENDIX II: Motor Comparison

Maxon RE 40

No load RPM	ω(max) rad/s	T(max) Nm	R	Angular Velocity rad/s	T(ext)
7580	793.775	2.5	50	0	125
7580	793.775	2.5	50	1	124.8425
7580	793.775	2.5	50	2	124.685
7580	793.775	2.5	50	3	124.5276
7580	793.775	2.5	50	4	124.3701
7580	793.775	2.5	50	5	124.2126
7580	793.775	2.5	100	0	250
7580	793.775	2.5	100	1	249.685
7580	793.775	2.5	100	2	249.3701
7580	793.775	2.5	100	3	249.0551
7580	793.775	2.5	100	4	248.7402
7580	793.775	2.5	125	5	310.5316
7580	793.775	2.5	125	0	312.5
7580	793.775	2.5	125	1	312.1063
7580	793.775	2.5	125	2	311.7126
7580	793.775	2.5	125	3	311.3189
7580	793.775	2.5	125	4	310.9252
7580	793.775	2.5	125	5	310.5316
7580	793.775	2.5	133	0	332.5
7580	793.775	2.5	133	1	332.0811
7580	793.775	2.5	133	2	331.6622
7580	793.775	2.5	133	3	331.2433
7580	793.775	2.5	133	4	330.8245
7580	793.775	2.5	133	5	330.4056

Mabuchi RZ 735

No load RI	ω(max) rad/s	T(max) Nm	R	Angular Velocity rad/s	T(ext) Nm
20400	2136	1.265	125	0	158.125
20400	2136	1.265	125	1	158.051
20400	2136	1.265	125	2	157.9769
20400	2136	1.265	125	3	157.9029
20400	2136	1.265	125	4	157.8289
20400	2136	1.265	125	5	157.7549
20400	2136	1.265	133	0	168.245
20400	2136	1.265	133	1	168.1662
20400	2136	1.265	133	2	168.0875
20400	2136	1.265	133	3	168.0087
20400	2136	1.265	133	4	167.9299
20400	2136	1.265	133	5	167.8512
20400	2136	1.265	150	0	189.75
20400	2136	1.265	150	1	189.6612
20400	2136	1.265	150	2	189.5723
20400	2136	1.265	150	3	189.4835
20400	2136	1.265	150	4	189.3947
20400	2136	1.265	150	5	189.3058
20400	2136	1.265	175	0	221.375
20400	2136	1.265	175	1	221.2714
20400	2136	1.265	175	2	221.1677
20400	2136	1.265	175	3	221.0641
20400	2136	1.265	175	4	220.9604
20400	2136	1.265	175	5	220.8568

A comparison between these two motors is given at different transmission ratios. Both of these can be used, but Maxon RE 40 provides better torque and velocity characteristics, which is why it is preferred.

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