Powered Ankle Prosthesis with Series-Elastic Actuation and Force Control

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Abstract

This work presents the mechanical design and control scheme of a powered ankle foot prosthesis platform with a Series-Elastic Actuator (SEA). The prosthesis has a DC motor in series with elastic elements that simulates the muscles and tendons in the human ankle and provides positive net to assist motion. Together with sensors, this prosthesis is capable of detecting external forces and react accordingly a variety of controllers. Three different controllers were tested separately, and within a finite state machine algorithm that simulates the ankle gait. When force sensing is incorporated in the controller, the prosthesis performs better compared to just a position controller. This will contribute to the future development of powered prosthesis that could be deployed in real-life situations such as walking in slopes, different terrains, and during star ascend and descend. Lastly, this platform aims to help robotic researchers to prototype controllers for powered prosthesis and other powered devices.

Introduction

In 1995 it was proposed by Pratt and Williamson to take advantage of the elasticity between the load and its actuator instead of trying to make it as stiff as possible[1], which was the current practice. The flexibility would allow for better control algorithms since sensors could be used to measure the offset from the expected actuator value. Pratt and Williamson proved that it was a viable way to introduce force control on the MIT humanoid robot Cog.

About ten years later, ankle-foot prosthesis were still passive, e.g. they used springs to store energy from the movement and release it at the right time. Au et. al. found a way to power the prosthesis and showed that it reduced the effort required from the clinical subject to operate it by between 7% to 20%[2]. This advancement led to the recent work by Eilenberg et. al., which introduces a way to use force feedback to make an adaptive controller for a similar prosthesis[3]. Up until then controllers had been specifically designed for a certain walking speed and terrain, but now the energy provided by the prosthesis was seen to be automatically adapted to the slope of the ground. They found that this new control method was producing results matching those of a human ankle, with differences that could be described by other reasons such as for example the clinical subject not being used to having a foot.

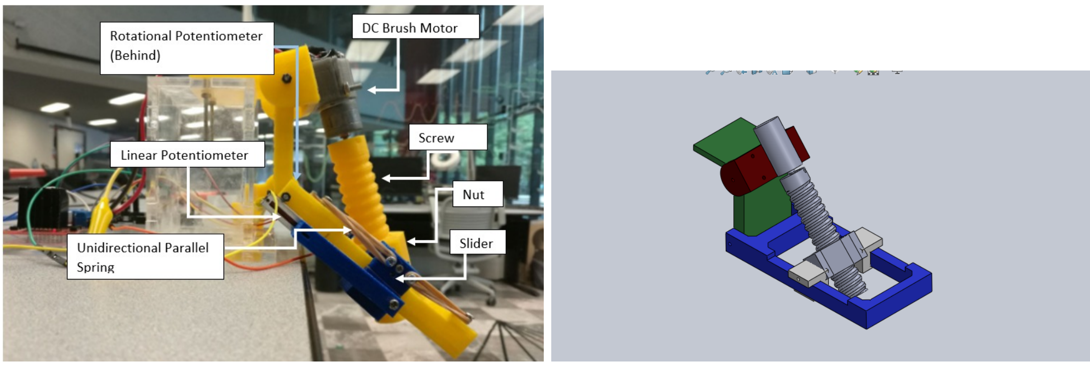
The advances made in laboratories have resulted in several commercial products; most notably are the exo-skeletons from ReWalk[4], and the prosthesis from Össur[5]. This state of the art technology aims to help people with different needs to have an as normal life as possible. The ReWalk exo-skeleton achieves this by providing a full body construction that aids people with walking. The prosthesis from Össur are smaller and replaces a specific limb, e.g. ankle or wrist, which behaves in the same way the missing limb did before.

By carefully building a prototype of a powered ankle-foot prosthesis, that allows for external forces acting on it to be measured, this project aims to develop control algorithms that can take action in regard to this input. This process is described in this report starting with a description of the prototype design and construction, system identification, and the design of different controllers, including a position controller, two force controllers, and a finite state machine. The next section includes the results from the experiments conducted and comparisons to similar work. In the final section the conclusions drawn by the authors are found which tries to reflect on the project, and what can be done in the future.

1 Work description

In this section, the technical information regarding the mechanical design and control algorithms is presented. First, the mechanical design is presented with an explanation of the pieces and the sensing hardware. Then, the methods to obtain the process dynamics and PI tuning are shown. Finally, an explanation of three controller designs is given.

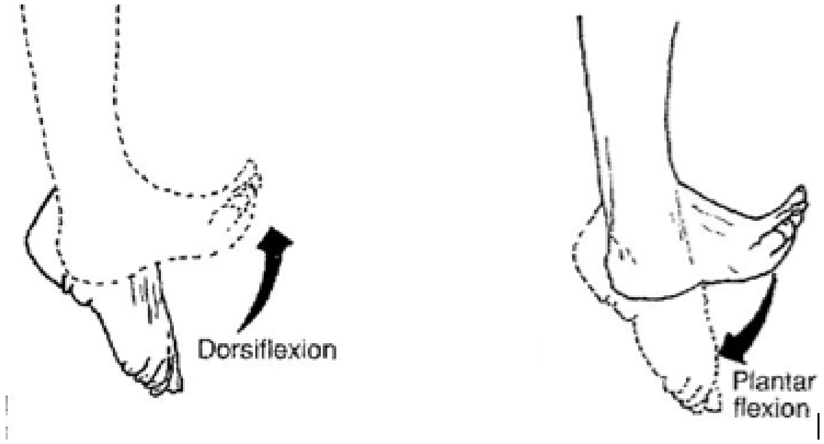
For the implementation of this project, a 3D prototype of a prosthetic foot was initially designed and assembled using Solidworks and was later printed, see fig. 1. The design consists of a DC motor (actuator) that actuates a turning bolt which takes the foot from its minimum to maximum positions by displacing a nut through a channel.



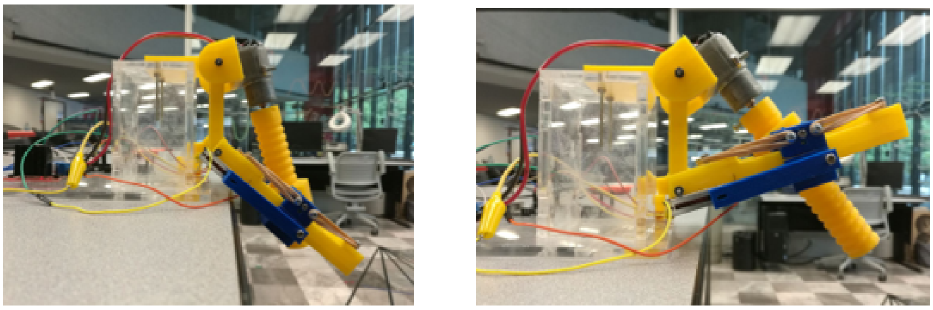
(a) Physical prototype (b) CAD model

Figure 1: Mechanical design of the prosthesis foot prototype.

Once the design was printed and assembled, before adding the sensing devices to get the foot's position, a pair of rubber bands were added on each corner of the 3.27 in $\times$ 7 in foot to hold the sliders that runs through the 4 in channel in position, this rubber bands act the way tendons would by holding the sliders. The motor on the other side has its own support (red Piece on fig. 1) this support is attached to the foot's base and it is able to move depending on the foot's position. To measure the foot's position, a rotational potentiometer was placed in the foot's joint. On the other side, a linear potentiometer was attached to the foot and the sliders to detect any external force to the foot. The whole mechanism simulates a biological ankle with the foot, being able to perform dorsiflexion and plantar flexion movements in the sagittal plane, see fig.2a and 2b.



(a) Ankle biological movement on the sagittal plane.



(b) Joint movement of the ankle prosthesis prototype.

Figure 2: Joint ankle movement.

1.1 Controller Design

1.1.1 Process Dynamics

To design a controller, the dynamics of the process need to be known. The first option to characterize our system was to measure the response of the DC motor when a step input of 4.8 volts was applied, and obtain the first order curve. This option was discarded because the team lacked of a tachometer or encoder to measure the RPMS of the DC motor. Instead, a ratio between the change in the voltage of the potentiometer that measures the foot position, and the time the step input of 4.8 volts lasted, was calculated. Ten experiments were done in each direction, dorsiflexion (going up) and plantar flexion (going down). The mean values were used to calculate the transfer functions.
Transfer function of dorsiflexion.

$$Y(s)/U(s) = 0.2894/s \tag{1}$$

Transfer function of plantar flexion.

$$Y(s)/U(s) = 0.2853/s \tag{2}$$

1.1.2 PID Tunning

Tunning the PID is a complex job, since there are many methods to choose from, for example, the Zieglers-Nichols method, or the ITAE. But we need to remember that the transitory response stability depends on the poles of the characteristic equation. For that reason, we created a theoretical characteristic equation with the desired parameters values and then equalize both of them; the desired equation with the equation with the parameters, and then solve for all the constants. It needs to be noted that both equations have to be the same order; so, if needed, multiply for a pole that is 10 times smaller than the smallest number (ten times to the left in the Cartesian plane).

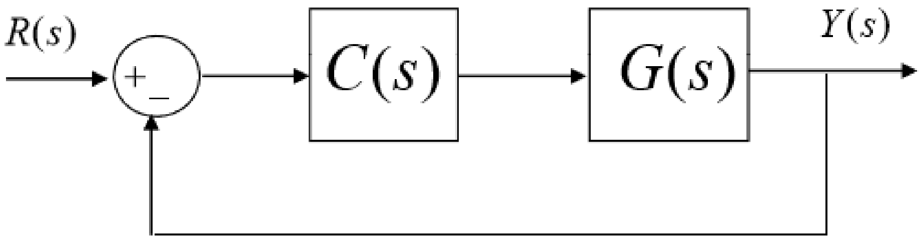


Figure 3: Closed loop diagram, where $R(s)$ is the reference, $C(s)$ the controller, $G(s)$ the plant model, and finally $Y(s)$ the output

Since we don't want an overshoot in the closed loop response we can use a first order control. For that, we use the equation 3. Where h_0 and h_1 are the coefficients of the polynomial to have the poles where desired.

$$\begin{aligned} H_t(s) &= h_0 s + h_1 \\ &\Rightarrow h_0 = 1 \end{aligned} \tag{3}$$

While equation 4 is the classical PI controller equation.

$$C(s) = K_p + \frac{K_i}{s} \tag{4}$$

Having these equations, and the control loop shown in fig. 3, we can solve for the transfer function of the model (eqn. 6).

$$G(S) = \frac{0.2844}{s} \tag{5}$$

$$T(s) = \frac{G(s)C(s)}{1 + G(s)C(s)} = \frac{0.2894(K_p s + K_i)}{s^2 + 0.2894K_p s + +0.2894K_i} \tag{6}$$

We want the process to stabilize in maximum 4 seconds, therefore from equation 3:

$$\begin{aligned} h1 &= \frac{1}{\tau} = \frac{4}{4} = 1 \\ h0 &= 1 \\ \Rightarrow H(s) &= s + 1 \end{aligned} \tag{7}$$

Since the characteristic equation of 6 is not the same order as eqn. 3, we need to multiply it by a pole 10 times smaller than the smallest, as shown in eqn. 8

$$H'(s) = (s+1)(s+10) = s^2 + 11s + 10 \tag{8}$$

Now we can equalize both equations(eqn. 8 and 3) and solve for the respective constants.

$$\begin{aligned} H_e(s) &= H_t(s) \\ s^2 + 11s + 10 &= s^2 + 0.2894K_p + 0.2894K_i \\ \Rightarrow k_p &= 38.00 \\ k_i &= 35.55 \end{aligned} \tag{9}$$

Finally substituting the values in eqn. 9 into eqn. 4 to obtain the following controller equation:

$$C(s) = 38 + \frac{35.55}{s}) \tag{10}$$

1.1.3 Position Controller

After the PID was tunned, the parameter *kd* was set to zero. A position controller was designed to evaluate the joint movement of our prototype. It is important to remark that further in the development there were made slight changes to the controller's constants in oder to optimize it. The functions used in this controller were the following:

$$e(k) = r(k) - c(k) \tag{11}$$

$$\begin{gathered} m(k) = (1/(2*T))*((2*T*m(k-2)) + ((2*T*Kp + Ki*(T^2) + 4*Kd)*e(k)) + \\ ((2*Ki*(T^2) - 8*Kd)*e(k-1)) + ((-2*T*Kp + Ki*(T^2) + 4*Kd)*e(k-2))) \end{gathered} \tag{12}$$

C(k) was the value of the potentiometer, indicating the joint position.
For this controller, two states were considered, one at the minimum and one maximum position of the ankle prosthesis. These values were calculated manually by reading the value of the potentiometer at each mechanical stop. The PI was used to reach the desired position in each state, and the algorithm runs continuously until the user indicates the stop condition.

1.1.4 Force Controller

Added to the position controller, the value of the linear potentiometer is considered in this algorithm. If no force is detected, the control system is just a position controller, if a external force is sensed, the value of the linear potentiometer indicates the direction of this external force. Two possible scenarios were programmed. In the first one, if an external force is detected, the prosthesis stops moving and only moves in the direction of the external force. In the second one, when an external force is detected, the prosthesis stops moving for some milliseconds, and then it continues with its original reference. These controllers aim to help the prosthesis to detect slopes, steps or terrain changes.

1.1.5 Finite State Machine for ankle gait simulation

With the objective of simulating the walking gait of the ankle, a finite state machine was developed to achieve this. The states proposed by [2] were included in our algorithm. These states can bee sen in fig. 4).

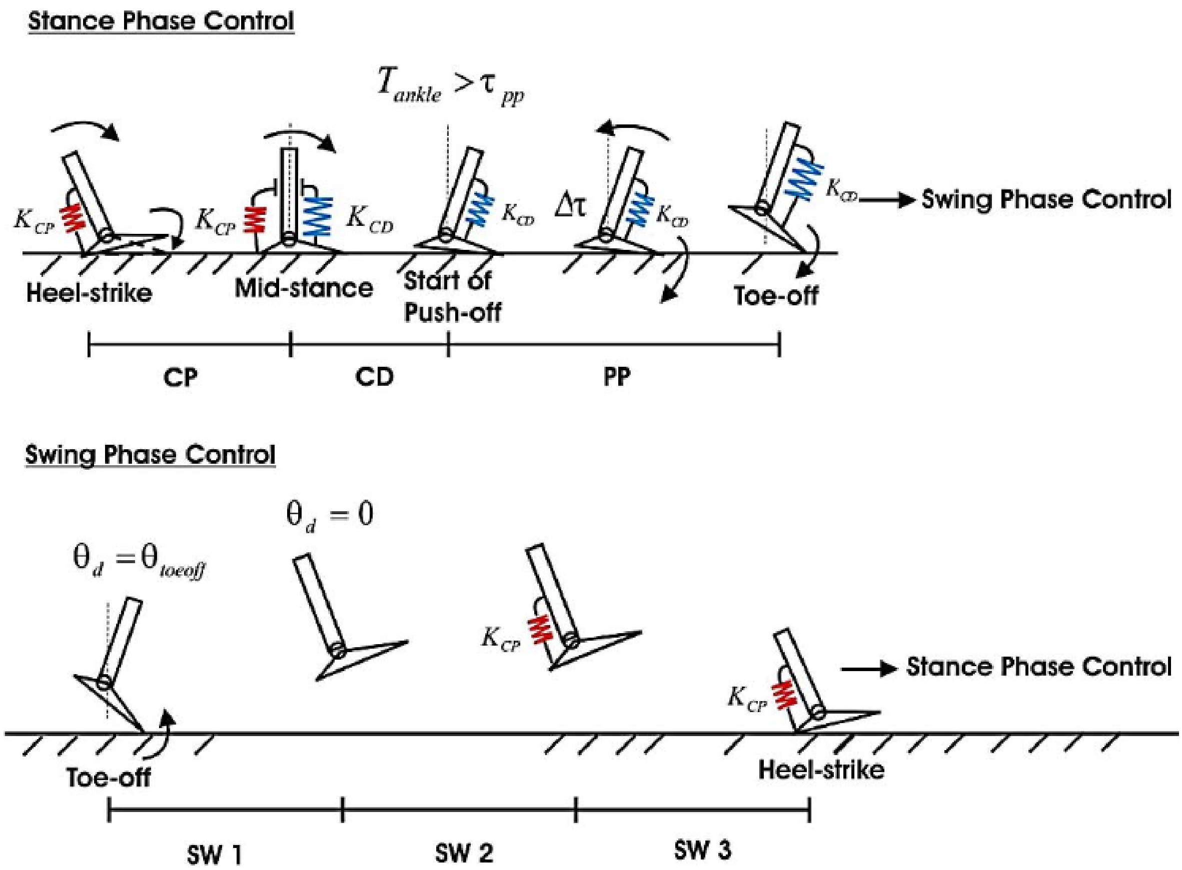


Figure 4: States for the walking gait cycle of an ankle joint.

CP begins at heel strike and ends when the foot is flat. CD begins at foot-flat and continues until the ankle reaches a state o maximum dorsiflexion. PP begins after CD and ends at the instant of toe-off. Finally, SW begins at toe-off and ends at heel strike.

2 Experiments

Several experiments were conducted to test the different controllers under certain conditions. A Graphical User Interface (GUI) was developed to allow visualization and storing of the experiments. Also, the position and force controllers were tested and compared between them.

2.1 Graphical User Interface

The GUI consists of a general panel with two graphing locations and several buttons to play run commands, as well as two sliders to adjust the set point and the manipulation of the process. The main purpose of the GUI is to visualize the status of the control signals in real time, however, it was integrated with tools to help integrate the main three control applications and test the position controller and the two force controllers. To improve the utility of the GUI, a switch to toggle between manual mode and automatic mode was integrated.

Figure 5: GUI main frame.

In figure 5 we can see in detail the main 5 control signals: the position output;*y(k)*, the force output;force, the position set point; *r(k)*, the position error; *e(k)*, and the actuator manipulation; *manipulation*. More into detail, in the figure we observe the reaction of the process to some pulses with the position control activated.

2.2 Position Controller implementation

The position controller was tested under two different conditions. First, it was tested with no perturbations to the foot, i.e, no slopes, steps or terrain changes, see the second plot in fig. 6. Then, perturbations or external forces were introduced, these can be seen in the first plot of fig. 6. When perturbations were present, it was expected to see changes in the position, since the foot could not move for a period of time due to the perturbation. However, for this controller, the prosthesis did not sensed these perturbations, hence the DC motor of the prosthesis continued moving during and after the perturbation was gone.

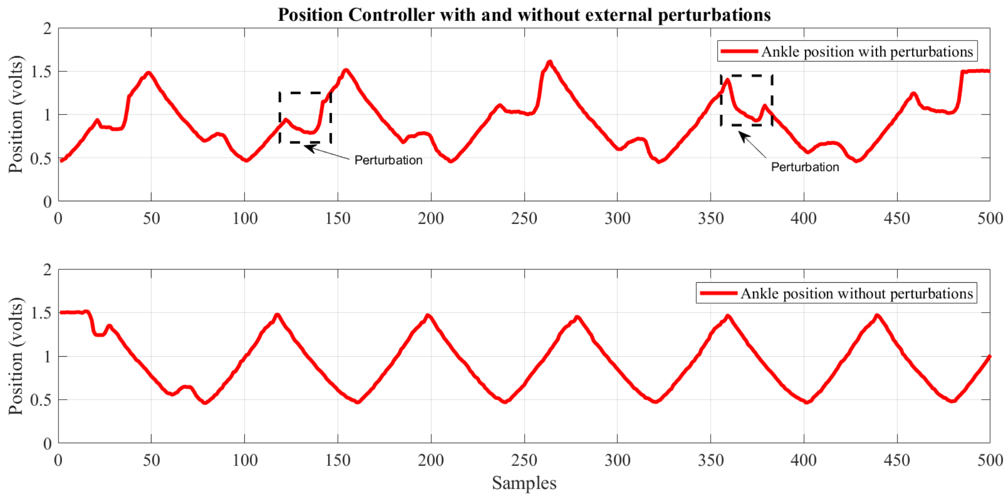


Figure 6: Position controller with and without perturbations. The X axis represents the samples the experiment lasted, the Y axis represents the voltage of the position sensing and the force sensing potentiometers.

2.3 Force Controller implementation

The force controller was tested in the two scenarios previously described in the controller design section. For the first scenario, if the prosthesis detected an external force, then it determined its direction according to the value of the linear potentiometer. If the external force its in the same direction, the prosthesis continued its movement, however, if it was in the opposite direction, it changed the reference and started moving in the direction of the external force. Position and force curves for this experiment can be seen in fig. 7. If no external force was detected, the controller behaved as a position controller following the PID reference. According to fig. 7,

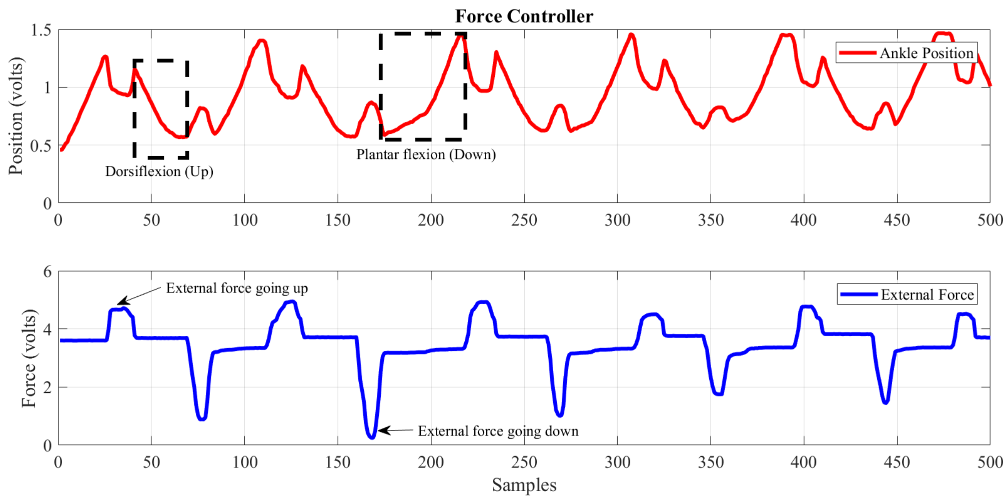


Figure 7: Force controller. The X axis represents the samples the experiment lasted, the Y axis represents the voltage of the position sensing and force sensing potentiometers.

it can be seen that when an external force is detected, the prosthesis starts moving according to the direction of this force, no matter the original trajectory of the prosthesis. If the external force is going up, the prosthesis performs dorsiflexion. If the external force is going down. the prosthesis performs plantar flexion.

For the second scenario, the controller did the following: if no external force was detected, the controller was a position controller following the reference according to the PID. If a external force or perturbation was detected, the prosthesis stopped its movement for some milliseconds, and continued with its original reference, until this reference was achieved. The curves for the force and position under this controller can be seen in fig. 8.

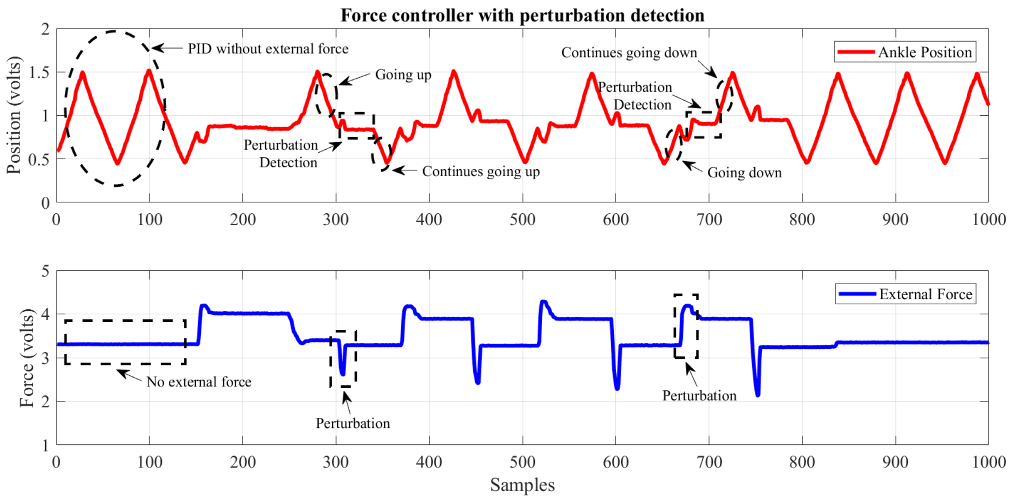


Figure 8: Force controller. The X axis represents the samples the experiment lasted, the Y axis represents the voltage of the position sensing potentiometer.

From fig. 8, it can be appreciated that when no external force is present, the position just follows its desired reference. When a perturbation to the prosthesis is present, the controller immediately detects it and stops moving, hence the position remains constant. After a certain time the prosthesis continues moving. This behavior is observed, either, during dorsiflexion or plantar flexion movements.

2.4 Finite State Machine implementation

To simulate the ankle gait, a finite state machine was implemented according to the states described in the previous section. During plantar flexion, CD and PP are performed. CD begins at heel strike and ends when the foot is flat, that can be seen in fig. 9, when the position values is around .77, PP starts immediately after CD and ends when the toe is off the ground, that is when the position value is around 1.4. During dorsiflexion, CP is performed, this state is equivalent to the swing states defined by [2], this occurs in the air, when the prosthesis is off the ground, and it is just to position the foot to be ready to perform the CD state at heel strike. The FSM implementation can be seen in fig. 9.

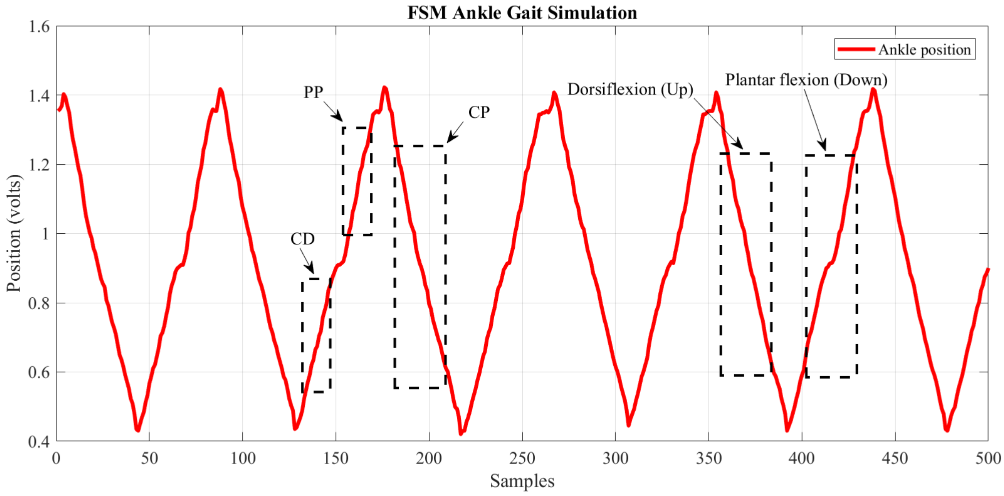


Figure 9: Ankle gait simulation on the prosthesis. The X axis represents the samples the experiment lasted, the Y axis represents the voltage of the position sensing potentiometer.

In fig. 9, the two states for plantar flexion; CD and PP are clearly distinguished, as well as the CP state for dorsiflexion.

3 Conclusion

In this project, a functional ankle powered-prosthesis prototype was built and tested with different controllers. The prosthesis consists of an actuator with a ball-screw transmission mechanism, and sensing hardware. The prototype also has rubber suspender in parallel to the actuator to simulate a series-elastic actuator (SEA). The SEA provides force control by controlling the extent to which the rubber suspenders are extended, following Hook's law. A position and two force controllers were designed and tested separately and in a finite state machine that simulates the ankle gait during walking. The results suggest that the prosthesis performs significantly better when the force is considered in the controller. This contributes to the design of powered robotic prosthesis, providing new insights for future controllers that consider changes in terrain, walking in slopes, star ascending, and descending, improving this technology and enhancing the quality of life of amputees. For future work, a more realistic prototype needs to be constructed in order to be tested with amputees in clinical settings. However, this project allows researchers and hobbyists to test controllers before they are tested in a real setting.

References

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