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Optimizing FES to Prevent Foot Drop

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Abstract
In this paper, we investigated the optimization of Functional Electrical Stimulation (FES) parameters for a foot drop assistive device. The model developed included the tibialis anterior (TA) muscle, which was stimulated, as well as the ankle joint. The main goal of the model was to stimulate the TA muscle with different levels of muscle excitations and determine a stimulus which would result in the lowest excitation level, while allowing the ankle to maintain a toe clearance height comparable to healthy individuals (10.1mm). The mechanics of the skeletal system were modelled as rigid bodies connected by joints that have limited degrees of movement; for example, the ankle joint was modelled as a hinge joint (allows movement in only one axis). State-space equations were developed in a 2D space to model the activation level, absolute orientation of the foot, and absolute velocity of the foot. Equations related to torque and isometric force were included, among others. The outcomes of this study revealed the most ideal excitation stimulus, and these results are verified with literature as being optimal. This value was chosen due to it fulfilling the main objective of the study: to maintain a minimum toe clearance height of 10.1mm during swing phase while minimizing fatigue in the TA muscle after being stimulated. The significance of this study is that it revealed a novel and efficient approach to minimize muscular fatigue through the method of finding an optimal stimulus (excitation level) for the TA. This approach can be used in clinical settings to reduce discomfort and fatigue in patients. This will optimize the current FES treatment by allowing patients to be trained more frequently, leading to faster recovery.

Keywords: FES, foot drop, optimization, stimulus, fatigue

Introduction

Background of Clinical Problem

Foot drop is a common neurological problem that is defined as the weakness of ankle and toe dorsiflexion [1]. The dorsiflexor muscles, which include the tibialis anterior (TA), extensor digitorum longus and extensor hallucis longus, help lift the foot during the swing phase of gait described in Figure 1. Foot drop occurs when these muscles are weak, resulting in difficulty in lifting the foot. This causes the foot to drag on the ground during the swing phase, increasing one's risk of falling or tripping. This problem often results from damage to the peroneal nerve, which sends electrical impulses to the muscles that lift the front of the foot. It is also caused by dorsiflexor injuries, neuropathies, and lumbar radiculopathy. Often, individuals who experience foot drop increase their steppage gait as a coping mechanism. This means they raise their leg higher to prevent their toes from dragging across the ground.

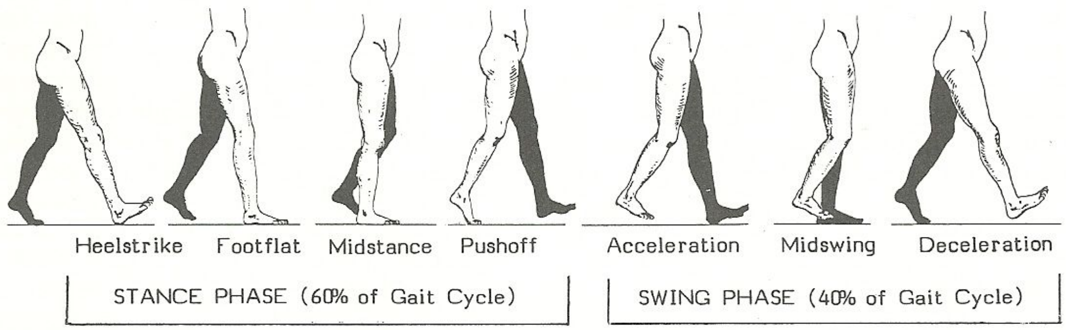

Figure 1: Diagram of phases during gait [2].

A common treatment to help control foot drop is Functional Electrical Stimulation (FES). This method consists of using a control box with electrodes which is worn under the knee [3]. This placement of electrodes is to allow stimulation of the peroneal nerves to send electrical impulses to the dorsiflexor muscles to lift the foot during gait. FES stimulates these nerves during the swing phase of gait shown in Figure 1; this occurs after push off and until the next heel strike. FES has been proven to allow those with foot drop to build more normal and comfortable walking patterns.

Project Goals/Objectives

The key objective of this project is to optimize FES parameters for a foot drop assistive device using a musculoskeletal model of the leg and foot. To accomplish this, state-space equations are developed to model the dynamics of the relevant muscles of the foot. Additionally, the muscle excitation and activation level are also modelled.

The purpose of the model is to determine the appropriate FES stimulus to trigger the dorsiflexion movement, thus preventing foot drop. The appropriate stimulus will be one that results in the lowest muscle fatigue and energy consumption by the TA muscle while still providing optimal foot clearance. It is important to lower the activation level of the muscle to reduce the fatigue it experiences under frequent stimulation. Since no formal optimization techniques are used, the minimum muscle excitation that prevents foot drop will be determined since, in this model, excitation directly influences activation level. This is sufficient because muscle fatigue is directly proportional to muscle activation over time; therefore, lower activation levels result in reduced fatigue [4].

There are some constraints that this model adheres to. First, the system modelled will be limited to the lower extremity (knee and below), specifically the TA and the ankle joint. Second, the foot rotation about the ankle must be caused by stimulating the TA; this constraint is added to ensure that the system models the functionality of existing FES devices. Third, the system will only be modelled during the swing phase of gait, which occurs during the latter 40% of the gait cycle [5]. Lastly, the toes must clear the ground with a height equal to or greater than the clearance height of healthy individuals, which is 10.1mm [6].

Relevant Prior Work

There are several prior models developed for the purpose of optimizing and improving Functional Electrical Stimulation (FES). One such model is described in the paper titled, “Nonlinear Model Predictive Control of Joint Ankle by Electrical Stimulation For Foot Drop Correction” [7]. The authors of this model will be hereinafter referred to as the literature group. This paper aimed to investigate the use of optimization control techniques to improve FES for foot drop correction. The model developed included the foot and TA muscle. The contraction of this muscle was controlled by electrical stimulation. The novel techniques used in this model involve modelling external states such as ankle accelerations, shank angle orientations, and angular velocity of the shank. These external states were measured via Inertial Measurement Units (IMUs) on hemiplegic patients; hemiplegia is the paralysis of one half of the body [7]. The control was optimized by minimizing the square of muscle excitations, which serves to accomplish the overall goal of reducing energy consumption in the muscle. First, an offline control problem was solved in order to determine the efficiency of FES optimal control for foot drop correction. Next, a Nonlinear Model Predictive Control (NMPC) was simulated to determine the feasibility of NMPC for foot drop correction. There were also some fixed constraints for foot orientation applied to the optimization problem. Then, an adaptive constraint on the ankle height was tested. The results of the fixed and adaptive constraints were compared. It was found that the adaptive constraint resulted in energy consumption that was three times lower than that of the fixed constraint. Therefore, the adaptive constraint established a novel approach to minimize energy consumption.

Motivation for Work

Foot drop is a neurological problem that is one of the most common walking problems experienced by individuals [1]. Those afflicted with this problem experience fatigue after walking short distances, expend greater energy, endure pain and are unable to walk quickly. FES treatments help individuals improve their walking speed, balance, and confidence as well as reduce fatigue and pain. Although FES provides many benefits, it also has some limitations. For example, since individuals are required to build muscle strength before FES can be implemented, it requires twice the amount of energy needed for normal walking, resulting in muscle fatigue which increases the risk of injury [7].

Thus, it is important to optimize FES parameters to reduce energy consumption of the dorsiflexor muscles in order to decrease the fatigue experienced by users. Decreasing muscle fatigue would allow users to participate in FES treatments more frequently and with less discomfort. This could lead to faster treatment and allow patients to have a more positive experience during treatment.

Methods

The two-dimensional representation of the overall system that was modelled is provided below.

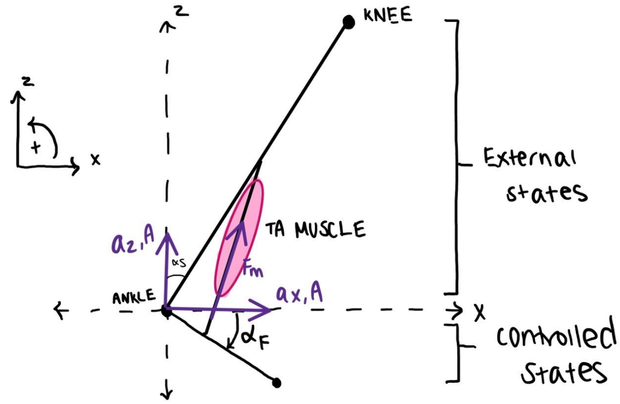

Figure 2: The model of the system from the knee to the foot.

In this model, the system is limited to the lower extremity (knee and below) and is represented in a two-dimensional space (X-Z). To model the mechanics of the skeletal system, it is assumed that they are rigid bodies connected by joints that have limited degrees of movement. The external states are used as inputs to the system and simulated against the state variables.

The state vector of the system is

$$\bar{x} = \left[x_1,\ x_2,\ x_3\right]^T = \left[f_{act},\ \alpha_F,\ \dot{\alpha}_F\right]^T \tag{1}$$

where f_{act} $(0 \leq f_{act} \leq 1)$ is the activation level of the muscle, α_F is the absolute orientation of the foot with respect to the horizontal axis, and $\dot{\alpha}_F$ is the foot's absolute rotational velocity [7]. All angles are positive in the counterclockwise direction. The input vector is

$$\bar{u} = \left[u_1\right] = \left[\varepsilon\right] \tag{2}$$

where ε $(0 \leq \varepsilon \leq 1)$ is the muscle excitation parameter. A u profile refers to the vector of different u values inputted to the model.

The system modelled is also dependent on external state variables which are defined in the following vector

$$\bar{x}^{ext} = \left[x_1^{ext},\ x_2^{ext},\ x_3^{ext},\ x_4^{ext}\right] = \left[a_{x,A},\ a_{z,x},\ \alpha_S,\ \dot{\alpha}_s\right] \tag{3}$$

where $a_{x,A}$ and $a_{z,x}$ are linear accelerations of the ankle with respect to the horizontal and vertical axes, respectively. α_s is the absolute orientation of the shank with respect to the vertical axis and $\dot{\alpha}_s$ is the absolute rotational velocity of the shank. These vectors are labelled as external since they are determined by parameters outside of the considered system. In literature, these values were measured from hemiplegic patients and recorded using IMUs for the accelerations and a gyroscope for the absolute orientation and velocity of the shank. A graph of the measurements for the shank orientation were given, however the accelerations and rotational velocities were not presented. Therefore, this model uses IMU measurements from a different piece of literature to get x and z acceleration data; these graphs are presented in Figure 3 and 4 respectively [8].

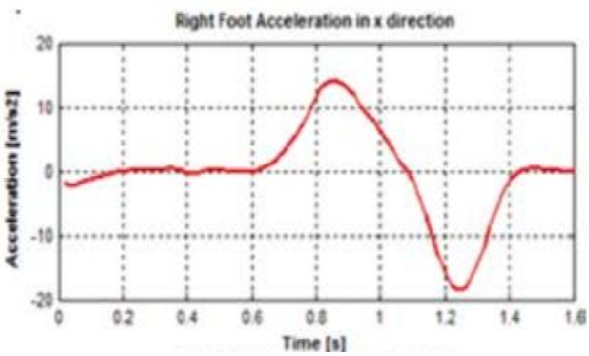

Figure 3: Acceleration in x-direction

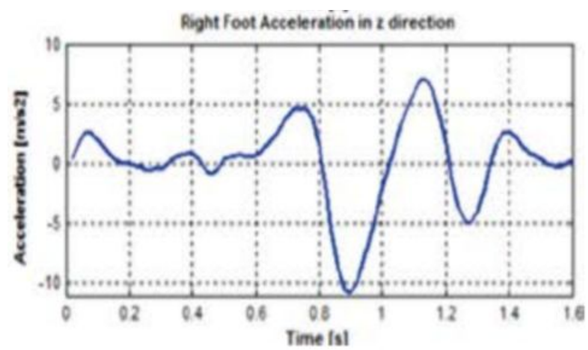

Figure 4: Acceleration in z-direction

These figures graph the range of accelerations recorded by an IMU throughout the duration of the gait cycle (0.6s to 1.4s). However, only the accelerations from the swing phase were needed, so the data was taken from the last 40% of the cycle (1.08s to 1.4s), terminating with heel strike [5]. Webplot Digitzer was used to gather the coordinates from the graphs. These data points were then horizontally stretched and shifted to fit the duration of 0 to 0.35 seconds. This time span was specifically required, as the paper that this model was based off of had all their simulations during this time period. Furthermore, the data for ankle height, one of the key values required for achieving our goals, was extracted from the paper that the model is based from and the data was provided for the duration of 0 to 0.35 seconds. Ankle height is not a variable in the model, but was externally measured in the literature [7].

The angular orientation of the shank throughout the gait phase given in the literature is presented in Figure 5, with *Toe off* and *Heel strike* signaling the start and end of the swing phase.

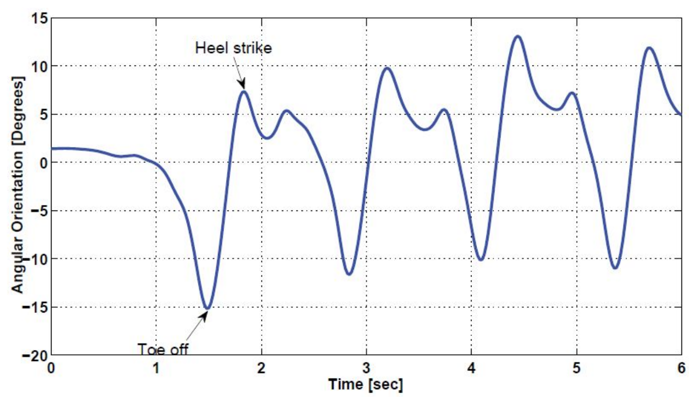

Figure 5: Absolute shank orientation obtained from literature [7]

Again, Webplot Digitzer was used to extract the data points during the swing phase. They were also time shifted, however very little time stretch was required, as this data spanned approximately 0.35 seconds in duration. These three external variables had their extracted data plotted and curve fit using numpy's polyfit function. This function has parameters for coordinates and the degree of the fitting polynomial. The degree parameter was found through trial and error while tracking the root mean square error (RMSE) which was normalized over the range of values. RMSE is the standard deviation of the residuals, which measures how far the data points are from the regression line; thus, RMSE reveals the degree to which data points are concentrated around the line of best fit. This can be used to determine the accuracy of the curve fitting. The equation used for the normalized RMSE (NRMSE) is provided in equation 4, where y_i is the data point and $\widehat{y}_i$ is the estimation by the fitted polynomial.

$$NMRSE = \frac{\sqrt{\frac{1}{n}\Sigma_{i=1}^{n}(y_i - \widehat{y}_i)^2}}{y_{max} - y_{min}} \tag{4}$$

Once an appropriate curve was determined for shank orientation, the derivative of the curve was calculated to get data for the absolute rotational velocity of the shank. No RMSE was calculated for shank velocity since its data was determined from the curve fit of shank orientation. There is no true data which can be compared against the calculated values, so RMSE calculations are not possible. All the data for these external state variables were extracted into *csv* files to be used later in the model. The main assumption made for this step was that the data from multiple sources would all work together correctly. The data time shift and stretch were essential in making sure that all of the variables changed together throughout the swing phase.

The state equations that describe the behaviour of the system are:

$$\dot{x} = \left[\dot{x}_1,\ \dot{x}_2,\ \dot{x}_3\right]^T \tag{5}$$

$$\dot{x}_1 = (u_1 - x_1)\left(\frac{u_1}{T_{act}} - \frac{1-u_1}{T_{deact}}\right) \tag{6}$$

$$\dot{x}_2 = x_3 \tag{7}$$

$$\dot{x}_3 = \frac{1}{J}\left(F_m(x_1, u_1)\,d + T_{grav}(x_2) + T_{acc}(x_1^{ext}, x_2^{ext}, x_2) + T_{Ela}(x_2)\right) \tag{8}$$

where T_{act} is the activation time constant, T_{deact} is the relaxation time constant and J is the inertia of the foot around the ankle. F_m is the force produced by the TA, d is the moment arm of the TA with respect to the ankle, T_{grav} is gravity torque of the foot around the ankle and T_{acc} is the torque produced by the movement of the ankle. The following equations were used to find the gravity torque and the torque produced by the ankle:

$$T_{grav} = -m_F \cdot c_F \cdot g \cdot cos(x_2) \tag{9}$$

$$T_{acc} = m_F c_F \left(x_1^{ext} sin(x_2) - x_2^{ext} cos(x_2)\right) \tag{10}$$

where m_F is the mass of the foot, c_F is the centre of mass with respect to the ankle, and g is the acceleration due to gravity.

T_{Ela} is passive elastic torque around the ankle due to different passive muscles and tissues; it depends on ankle and shank position. The equation that denotes the passive elastic torque around the ankle is:

$$T_{Ela} = exp(a_1 + a_2 x_2) - exp(a_3 + a_4 x_2) + a_5 \tag{11}$$

where the values of a_{1-5} are the parameters of elastic torque found from literature [7].

$$\left[a_1,\ a_2,\ a_3,\ a_4,\ a_5\right] = \left[2.10,\ -0.08,\ -7.97,\ 0.19,\ -1.79\right] \tag{12}$$

When the TA is electrically stimulated, it produces a force F_m, and controls the orientation of the foot relative to the shank. The force produced is defined by the following equation:

$$F_m = x_1 F_{max} f_{fl}(x_3^{ext} - x_2) f_{fv}(x_4^{ext} - x_3) \tag{13}$$

where F_{max} is the maximal isometric force, and f_{fl}, f_{fv} are non-linear force-length and force-velocity relationships, respectively. f_{fl} links the generated force to the length of the muscle, defined as follows:

$$f_{fl}(x_3^{ext} - x_2) = exp(-[-\frac{l_{CE}-l_{CE,opt}}{Wl_{CE,opt}}]^2) \tag{14}$$

where $l_{CE,opt}$ is the optimal length of the muscle fibres for force generation, W is the shape parameter defining a range of acceptable muscle displacements for force generation, and l_{CE} is the true length of the muscle. l_{CE} is calculated by finding the difference between the length of the muscle tendon complex (l_{MT}) and the tendon length (l_T), defined below:

$$l_{CE} = l_{MT} - l_T \tag{15}$$

l_{MT} is defined by the following equation:

$$l_{MT} = l_{MT,0} + d(x_3^{ext} - x_2) \tag{16}$$

where $l_{MT,0}$ is the muscle-tendon length at rest, and d is the moment arm of the TA with respect to the ankle.

The force-velocity relationship f_{fv} is defined as follows:

$$\begin{cases} f_{fv}(x_4^{ext} - x_3) = \dfrac{1-\frac{v_{CE}}{v_{max}}}{1+\frac{v_{CE}}{v_{max}f_{v1}}} & \text{if } V_{CE} < 0 \\ f_{fv}(x_4^{ext} - x_3) = \dfrac{1+a_v\frac{v_{CE}}{f_{v2}}}{1+\frac{v_{CE}}{f_{v2}}} & \text{Otherwise} \end{cases} \tag{17}$$

where v_{max} is the maximal contraction speed, and a_v, f_{v1}, and f_{v2} are force-velocity parameters. V_{CE} is the contraction speed of the muscle, equal to $d(x_4^{ext} - x_3)$.

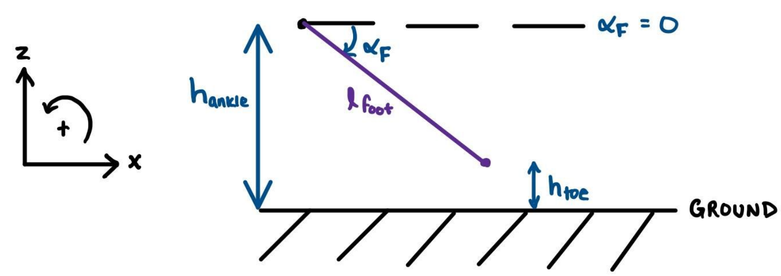

Figure 6: Diagram of the ankle and toe height.

From Figure 6, the equation for the height of the toe from the ground can be calculated as follows:

$$h_{toe} = h_{ankle} + sin(x_2) \times l_{foot} \tag{18}$$

where h_{toe} is the height of the toe from the ground, h_{ankle} is the height of the ankle from the ground, x_2 is the second state variable α_F, the absolute orientation of the foot with respect to the horizontal axis, and l_{foot} is the length of the foot, equal to 0.23 metres.

The chosen starting state was slightly different than that of the paper, because h_{toe} needed to start at zero. To determine the starting angle, equation 18 was solved for x_2, resulting in a value of -45 degrees from the horizontal. This value is higher than normal since it is measured from someone with foot drop, where a common coping mechanism is raising the leg and foot higher than normal [1]. This is the only significant difference between our model and the papers [7]. All of the provided equations were coded in Python. The solve_ivp function was utilized to simulate the model between 0 and 0.35 seconds. After the initial plots were made, sanity checks were conducted.

The first sanity check was with a zero *u* profile (no input stimulation).

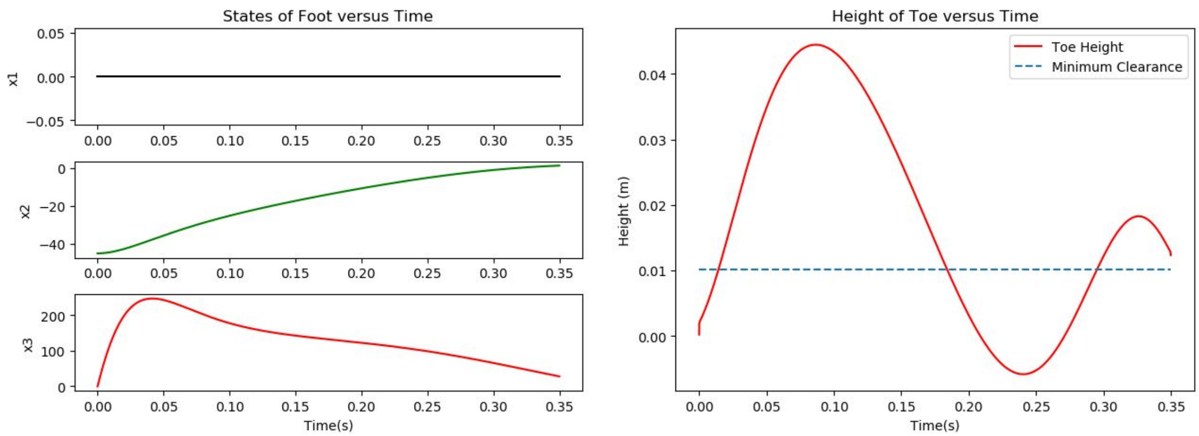

Figure 7: Sanity check for no excitation.

Although the toe height becomes negative in this simulation, it makes physical sense. Without any stimulation, the foot hangs limply from the ankle. This means that any rotational velocity or angle change is due to external variables (such as shank rotation). In this case, the person trips since their foot fails to clear the floor by a safe margin. In real life their foot would not actually go into the ground, but this model does not cover that.

The second sanity check was conducted with a constant maximum stimulation. This would mean that the foot would rapidly reach its maximum dorsiflexion angle (20 degrees) [9].

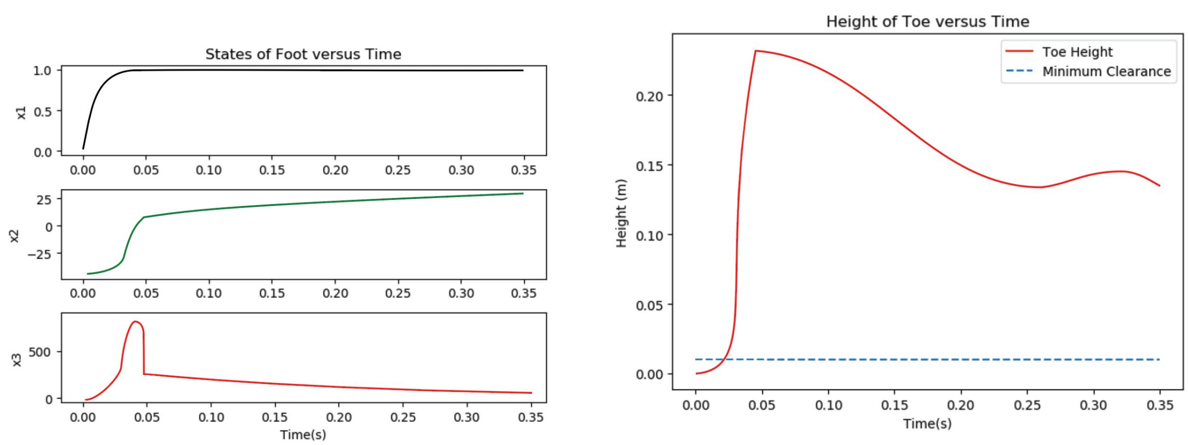

Figure 8: Sanity check for full excitation.

This is not a natural motion which is why the graphs seem nonsensical, but upon inspection, the graphs follow the predictions of the model. The x_2 graph increases rapidly until it has a kink, which is where the foot reaches its maximum dorsiflexion. The rest of the graph is smooth since the angle only changes due to the external state variables. The x_3 graph also increases very quickly, to reach a value almost 3 times as high as the no stimulation sanity check.

The rotational velocity of the foot is much greater since the angle of the foot drastically changes within 40ms. The toe height graph also has a very steep increase, until it also hits a kink, at which point it follows the curve of the ankle height. The values for toe height seem extremely high, but they make sense since the ankle height reaches a maximum of approximately 0.19m from the ground. Since the foot is already at a positive angle from the ground at this point, the toe height is higher than the ankle height. The final toe height is very high, but it makes sense since the foot is greater than a 30 degree angle from the horizontal.

There are three different *u* profiles that are going to be simulated using this model. The first one is a uniform activation, like the second sanity check. The results from this test should show what happens when the muscle is activated early on and stays activated for the duration. The second *u* profile is an exponentially increasing excitation. The reasoning behind this profile is that less excitation is needed initially to clear the toe height requirement, but the TA needs more activation as the foot begins to come down for heel strike. The third *u* profile is based on the ideal excitation pattern presented by the literature group [7]. The literature group utilized optimization techniques to achieve this pattern, so it is expected to perform the best. Our *u* profile varies slightly from the literature, but follows the general trend. There were other *u* profiles tested but they were ineffective and not mentioned.

Following this, each simulation was judged against a set of criteria. This aimed to ensure that the simulation was effective in preventing foot drop. As mentioned, the main criteria used to determine whether a given excitation is *successful* is if it ensures that the foot has a clearance of at least 10.1mm during the swing phase. The toes start on the ground, so the toe height will not always be above the minimum clearance. To ensure that this metric is still useful, toe height after 0.05s will be compared to the minimum clearance. This parameter is treated as a simple pass-fail, as failure to abide results in the tripping of the user, and unacceptable result. Since this can be achieved by a variety of excitations, the next important differentiator is muscle fatigue. To evaluate fatigue, no formal calculation was used. Fatigue was compared between the simulation as the integral of muscle activation (x_1) over time. This is based on the initial assumption that muscle activation is directly related to fatigue [4]. This metric was used for the simulations that passed the initial toe clearance requirement since the least resulting fatigue implies the greatest comfort of the user. The integral of u was determined to be the energy used by the FES device to provide the stimulation to the TA. Thus, the simulation that consumed the minimum amount of energy, while having low muscle fatigue, was chosen as the optimal simulation type.

Results
In this study, the optimal stimulation values for the prevention of foot drop were investigated. The outcomes of the methods previously described are depicted below. Before inputting different stimulus values, curve fitting was performed for all data extracted from external sources to obtain the equations of lines of best fit for the points. The most important data extracted is ankle height, since it is used to calculate toe height (the main performance metric); which is displayed in Figure 9. The original extracted data as well as the line of best fit found through numpy's polyfit method are depicted.

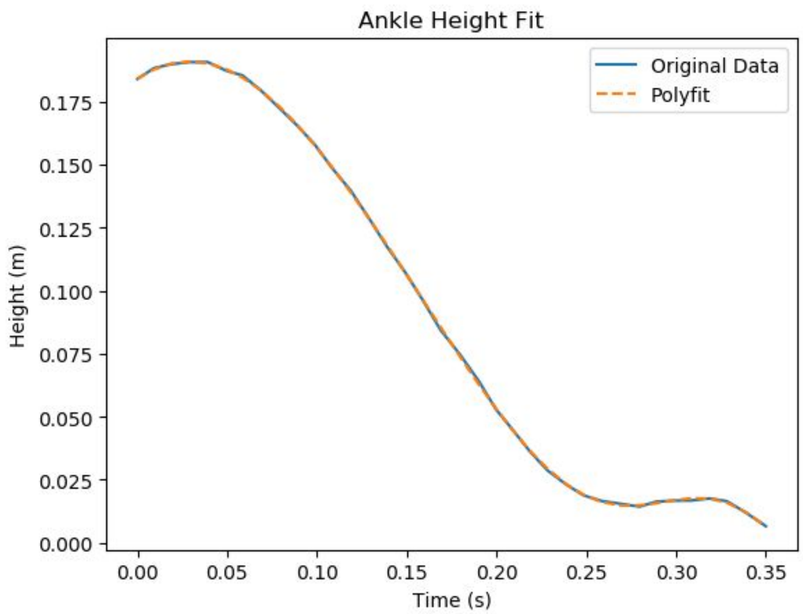

Figure 9: Curve fitting for ankle height data.

The slight increase in ankle height at the end can be attributed to IMU sensitivity error. Using physical measurement devices such as IMUs can lead to these errors due to their sensitivity to noise (such as vibrations or strong air flow) from the external environment.

Table 1: NRMSE values for data.

Data Type	NRMSE
Ankle Height	0.00285
Shank Angle	0.00518
X Acceleration	0.01602
Z Acceleration	0.02292

The NRMSE values calculated for all the external data, displayed in Table 1, help to define the errors that occur due to curve fitting. It was found that the curve fitting of the Ankle Height provided the lowest NRMSE with a value of approximately 0.00285. The Shank Angle curve fit resulted in an NRMSE of 0.00518, the X Acceleration curve fit resulted in an NRMSE of 0.01602 and the Z Acceleration curve fit resulted in the highest NRMSE with a value of

0.02292. All these errors are less than 0.05 which reveals that the curve fits were accurate to the original data points extracted. After the accuracy of the curve fits was confirmed, the simulations were performed.

In the first test, the effect of a uniform muscle excitation level on the toe height was investigated and is depicted in Figure 10.

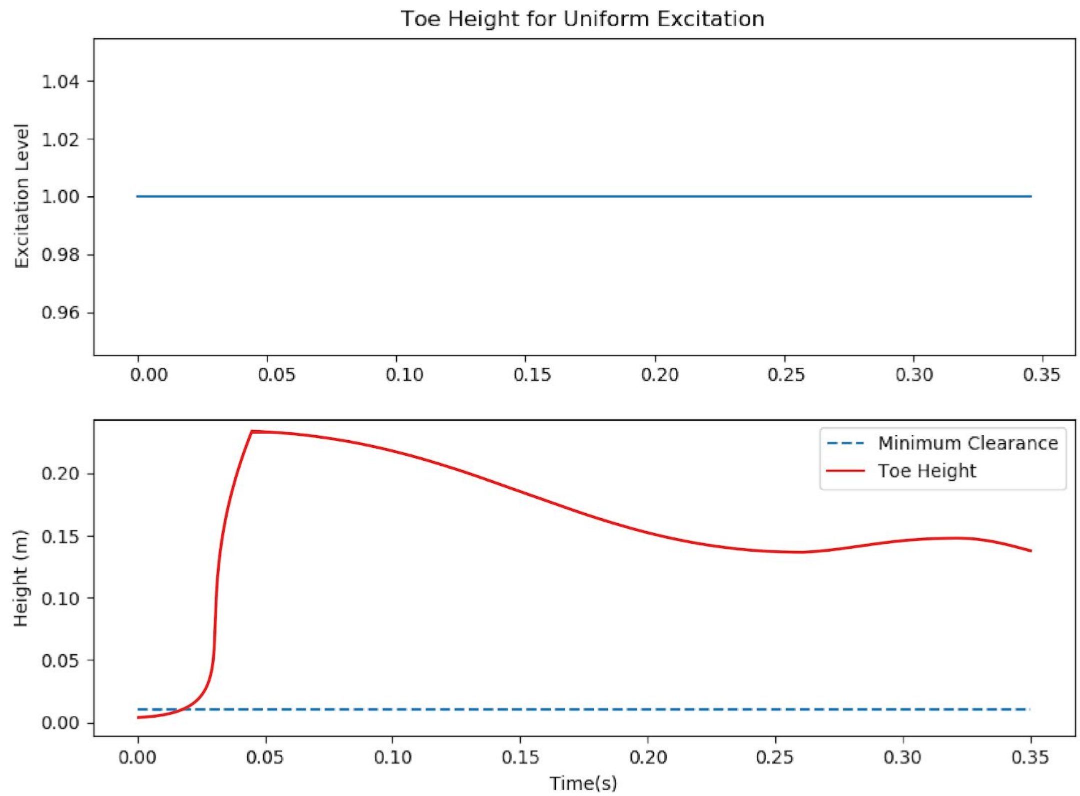

Figure 10: Toe height for uniform excitation.

When excitation is kept constant throughout the duration of the swing phase, the toe height increases rapidly at first and then slowly decreases to approximately 0.13m. As mentioned, this is due to the foot reaching its maximum dorsiflexion angle and then matching the motion of the ankle for the remainder of the motion. This excitation exceeds the minimum toe clearance requirement significantly, so it is not likely to be the most efficient excitation.

In the second test, an exponential excitation parameter was inputted and its effect on toe height was investigated. The results of this test are displayed in Figure 11.

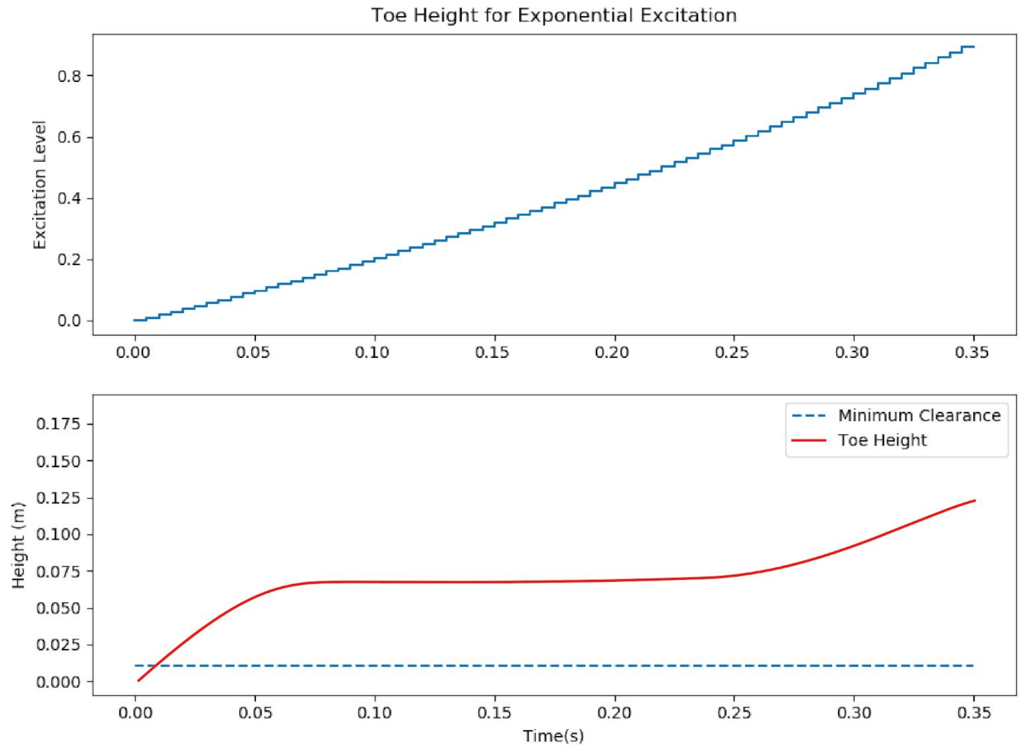

Figure 11: Toe height for exponential excitation.

When excitation is exponentially increased throughout the duration of the swing phase, the toe height increases slowly to approximately 0.06m, slightly increases for about 0.2 seconds and then increases slowly to approximately 0.125m. Unlike the uniform excitation, this excitation profile has a more gradual toe height increase. It also achieves our toe clearance height requirements, but the final toe height is almost as high as the uniform excitation. The plateau in the middle is caused by increased TA activation while the ankle height is decreasing proportionally. This sharper increase at the end is due to increasing excitation which leads to further activation and greater dorsiflexion. This method is also not likely to be the most effective.

In the third test, a muscle excitation level found to be ideal from literature was inputted to the model to reveal its effect on toe height. The literature excitation is provided in figure 12.

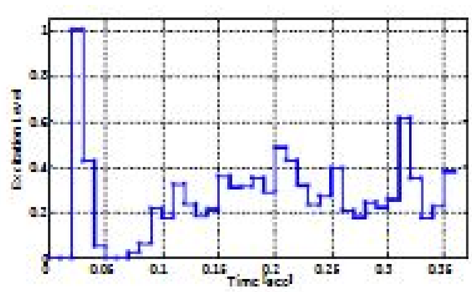

Figure 12: Ideal excitation from literature [7].

When this ideal excitation is used to stimulate the muscle during the swing phase, the toe height increases rapidly at first to approximately 0.06m and slowly decreases throughout the rest of the phase (its lowest being at about 0.03m). The simulation results are depicted in Figure 13.

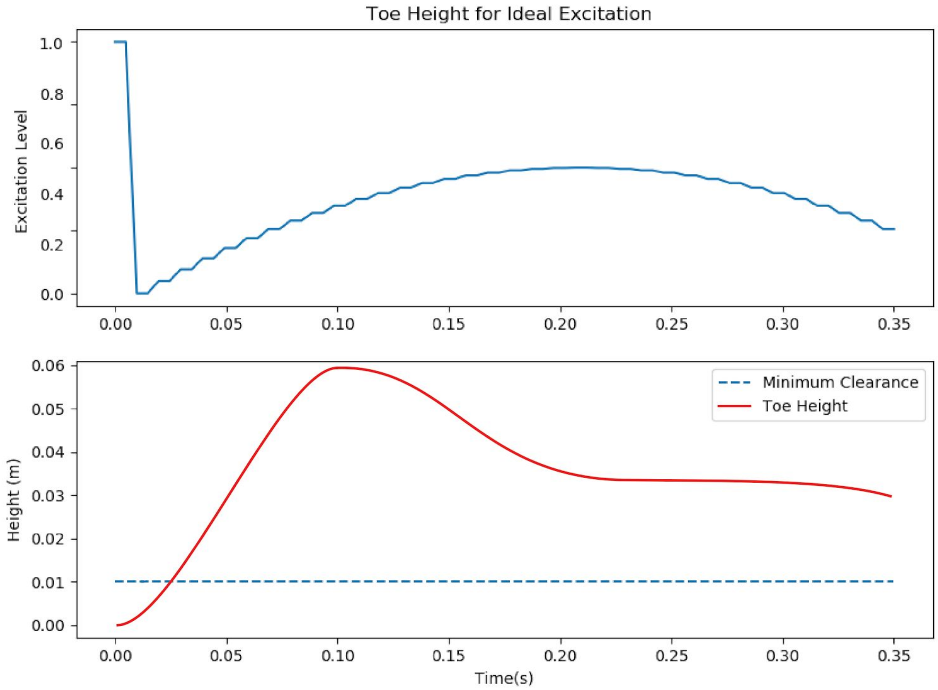

Figure 13: Toe height for ideal excitation.

This initial increase is due to the initial excitation, but it does not exceed the height from either the uniform or exponential excitations, while still ensuring the toe clearance is met. The decrease in the height is due to ankle height decrease with activation that is not as strong as the others. The toe height drops, but is still above the minimum clearance. The final toe height is notably lower than the other simulations. These results are similar to that of the literature group. They kept a greater clearance of 30mm, which appears to still be satisfied in this simulation.

Table 2: Integral calculations of the simulation results.

Simulation Type	Integral of Excitation Level	Muscle Fatigue (Integral of x_1)
Uniform Excitation	0.35000	0.32678
Exponential Excitation	0.13988	0.18591
Ideal Excitation	0.12523	0.14832

Table 2 displays the integral of the excitation profile (u) and the integral of its corresponding activation level (x_1) graph for each simulation performed. The integral of the excitation parameter relates to the energy consumed by the FES device to excite the TA of the user. The integral of the activation level is a representation of muscle fatigue as it is directly proportional to muscle activation over a period of time [4]. The uniform excitation resulted in the highest integral of excitation level (0.3500) and the highest muscle fatigue (0.32678). The exponential excitation stimulus resulted in the second highest integral of excitation level (0.13988) and the highest muscle fatigue (0.18591). Lastly, the ideal excitation stimulus resulted in the lowest integral value for excitation level (0.12523) and the lowest muscle fatigue (0.14832). These results match the expected outcome determined by the group.

Discussion

Summary of outcomes

Functional electrical stimulation of the foot was modelled against varying levels of muscle excitation, on the TA muscle, in order to determine an optimal stimulus that can prevent foot drop during a normal walking cycle. The main objective of this simulation was to determine which type of stimulus of muscle excitation would result in the lowest energy consumption of the TA muscle, while still maintaining a toe clearance height of 10.1mm [6], to eliminate tripping due to foot drop during swing phase. Various stimulus trends were modelled for this study, however the three plots outlined in the previous section display the worst (no excitation, Figure 10), medial (exponential excitation, Figure 11), and the best (optimized excitation, Figure 13). The most ideal stimulus was decided upon based on whether the two initial requirements were met, not crossing the minimum toe clearance height (10.1mm), and providing the least amount of fatigue on the TA muscle. This was calculated by integrating the muscle excitation and activation curves, as energy consumption and muscle fatigue [4].

The outcomes for this study revealed the optimal input for the excitation stimulus as the ideal excitation from the literature, shown in Figures 12 and 13. The literature group determined this excitation through formal optimization approaches, so it makes sense that it produced the best results [7]. As seen from this graph, the result of using this stimulus allows the toe to be at an acceptable clearance height, by not crossing near the minimum, while experiencing the least amount of fatigue on the TA muscle, as listed in Table 2. This follows the initial goals and objectives of the study.

Relating to the clinical problem

The results relate to the clinical problem of interest being to prevent foot drop from occurring during the normal walking cycle. By uncovering the ideal level of activation for the TA muscle, it can be used as a parameter to optimize foot drop assistive devices for those that experience this gait abnormality. It was very important to ensure that this ideal stimulus not only prevents the individual from crossing the minimum toe height from the ground, but also limits the amount of fatigue experienced by their TA muscle. The drop foot syndrome prevents the individual from performing dorsiflexion during the swing phase of their walking cycle and as a consequence forces them to adapt their walking gait to avoid dragging their toe on the ground. This makes one involve their hip and knee in excessive flexion, which leads to an inefficient gait. FES can be used to restore the functional behavior of paralyzed limbs, however can induce muscular fatigue and therefore increases the risk of injury if one is not able to adapt to its behaviour [7]. The results of this study can help optimize the parameters of FES so that a more efficient method can be implemented into the assistive-devices to allow individuals to adjust their gait in a method better optimized for them and their walking motion.

Limitations

There are some limitations that may restrict the extent of this model's accuracy. One, for example, would be that the simulation was entirely run in 2 dimensions, that being in the X-Z planes. This was done as the skeletal system was assumed to follow the mechanics of rigid bodies that are connected by joints and have limited degrees of movement. While this made it simpler to model the FES, realistically it would not be accurate as this negates all measurements in the Y direction. This removal of data can render the results being inaccurate as the muscles are only contributing to the X and Z movements of the body. Their contributions to the third dimension may lead to fatigue or other motor activity, which would not be displayed using this simulation. As well, many angular measurements were taken as inputs for this system, and since the Y plane was not included, any inversion or eversion foot was not considered. This makes the model slightly unrealistic as during the normal walking cycle, one would experience changes in their foot or ankle axial orientations, which would not be taken into account for this study.

Recommendations

Recommendations can be made to improve the accuracy and precision of this study. One example would be to generate new data that can be used as the external states for the system, as opposed to using data from external literature. For the purposes of this study, and due to time constraints, data regarding the linear acceleration of the foot in the X and Z directions, the orientation of the shank from the vertical axis, and the rotational velocity of the shank were taken from different pre-existing studies. This was useful for this simulation as in order to model the swing phase accordingly, these variables needed inputs changed with time. The issue with using external data was that they were found from different sources. While the data collected is still accurate, their combination limits the precision when being passed into the simulation. This was done since the times did not match up together, as they were taken from different pieces of literature, and the values were fit to match the same period throughout. One method to mitigate this would be to independently measure the external states through experimentation using an IMU or gyroscope attached to someone undergoing a swing phase. This would ensure each piece of data matches up with the same time period as they would all be recorded together. Multiple trials can be done as well, to improve the accuracy of the data as it seems fit.

Another recommendation for this study, would be to model the system in 3-dimensions. As of right now, the skeletal system of the foot and ankle is assumed to be in 2-dimensions, specifically the X-Z plane. While negating the Y data made it simpler to model, it made it more unrealistic. By integrating the Y-plane into the system, a further investigation of foot dynamics and structural responses during motor activity can be done [10]. This would make the model more accurate and fit to realistic data, which would be used when actually designing an assistive foot drop device. Although there would be a greater number of parameters and assumptions to consider, along with an entirely newer different skeletal model to work with, generating a simulation to fit a 3-dimensional model would be a great addition to further this study. This can be done in the future by again using an IMU or gyroscope to get all related acceleration and angle measurements in all three planes in order to collect all the related data.

Lastly, another recommendation for this study would be that the initial value for the state x_2 is currently lower than it should realistically be for someone who experiences normal gait. As explained, this variable was calculated to be -45 degrees in order to keep the initial toe height from the ground at 0. This measurement was used as the ankle height data collected was by someone who experiences foot drop, and hence resulted in a lower starting angle. This does limit the accuracy of the simulation, and is recommended that for the future in order to mimic normal gait, and optimize parameters of FES, data should be collected from healthy individuals. This way the excitation can result in a normal gait, rather than fixing an erroneous one.

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Appendix

Table 1 - Model Parameters and their Values

Parameters	Signification	Values [unit]
T_{act}	Activation constant time	0.01 [sec]
T_{deact}	Relaxation constant time	0.04 [sec]
J	Inertia of the foot around ankle	0.0197 $[kg.m^2]$
d	Moment arm of TA w.r.t the ankle	3.7 $[cm]$
B	Viscosity parameters	0.82
c_F	COM location w.r.t the ankle	11.45 $[cm]$
m_F	Mass of the foot	1.0275 $[Kg]$
a_v	First force-velocity parameter	1.33
f_{v1}	Second force-velocity parameter	0.18
f_{v2}	Third force-velocity parameter	0.023
v_max	Maximal contraction speed (shortening)	-0.9 $[m/\sec]$
F_{max}	Maximal isometric force	600 $[N]$
W	Shape parameter of f_{fl}	0.56
l_T	Constant tendon length	22.3 $[cm]$
$l_{MT,0}$	Muscle-tendon length at rest	32.1 $[cm]$
$\begin{pmatrix} a_1 \\ a_2 \\ a_3 \\ a_4 \\ a_5 \end{pmatrix}$	Parameters of elastic torque T_{Ela}	$\begin{pmatrix} 2.10 \\ -0.08 \\ -7.97 \\ 0.19 \\ -1.79 \end{pmatrix}$