Optimizing Functional Electrical Stimulation Parameters for Foot Drop Treatment

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Abstract

Foot drop occurs when an individual's tibialis anterior muscle does not function properly and can no longer adequately dorsiflex the foot to clear the ground prior to the heel strike phase of the gait cycle. Although Functional Electrical Stimulation (FES) is the preferred solution for foot drop, it overstimulates the targeted muscles, leading to muscle fatigue. The goal of this project is to determine the optimal parameters for FES. Optimal parameters would successfully reduce muscle fatigue caused by FES while lifting the foot sufficiently off the ground to prevent foot drop. One model was created and 900 simulations were run to determine the optimal amplitude and frequency of the FES signal. A simplified model of the leg, shank, and foot was used to investigate the relationship of the height of the foot off the ground with frequency and amplitude of the input signal. An exhaustive search determined that the optimal frequency and amplitude to be 50Hz and 2.177V, respectively. A final simulation was run using the simplified model of the leg, shank and foot to validate the results from the previous simulations. All simulations involved the tibialis anterior, soleus, quadriceps, and hamstrings muscle as well as the knee and ankle angles in a state-space model to simulate the leg behaviour over the course of the swing phase of the gait cycle. The results found that the given parameters produced realistic foot and ankle behaviour and ensured that both the front and back of the foot cleared the ground, preventing foot drop. Future iterations will consider the movement of the leg at the hip and use exact measurements to improve the model's accuracy and potentially make individualized recommendations for different individuals.

1 INTRODUCTION

Foot drop affects an individual's gait pattern and affects both the stance and swing phase (Carolus et al., 2019). The swing phase comprises 40% of the gait cycle, and can be subdivided into the initial swing, mid swing and terminal swing phases (DeLisa, 1998). During the initial swing, the tibialis anterior is primarily responsible for dorsiflexion of the ankle (DeLisa, 1998). Foot drop occurs during the swing phase when the tibialis anterior is unable to adequately dorsiflex the foot in order to clear the ground before the heel strike. Foot drop is indicative of a neurological, muscular or anatomical problem (Mayo Clinic, 2020). Currently treatments for foot drop include ankle foot orthoses (AFO) and functional electrical stimulation (FES). Although AFOs are successful in improving walking speed, gait pattern, and energy expenditure, many patients prefer FES (Bulley et al., 2011). FES elicits a more natural walking pattern when compared to AFOs although it is associated with rapid muscle fatigue which hinders its ability to continuously and effectively stimulate the muscles involved in dorsiflexion of the ankle to avoid foot drop (Thrasher et al., 2005).

Previous studies have investigated the effects of varying three FES parameters: pulse width, amplitude and frequency (Cheng et al., 2004). The study by Sabut et al. (2010) investigated FES parameter ranges of 10-50 Hz for frequency, 10-500 μs for pulse width, and 0-120 V for amplitude to stimulate the peroneal nerve (Sabut et al., 2010). Many current FES devices deliver the signal using a ramp-up pattern after the heel lift and ramp-down at the heel strike, although studies have investigated alternative delivery methods and have achieved a more natural gait pattern (Sabut et al., 2010). Optimal FES parameters are established when the tibialis anterior is stimulated as naturally as possible, gait is improved and muscle fatigue and pain are reduced (Sabut et al., 2010). Many studies have used a modified Hill muscle model to study the effects of FES parameters in the absence of human subjects. The models take input parameters including amplitude, frequency and pulse width, and produce joint torque as the output (Ferrarin and Pedotti, 2000).

The goal of this project was to determine the optimal FES parameters that successfully reduce muscle fatigue caused by FES while still adequately lifting the foot off the ground to prevent foot drop. The model was created using MATLAB and considered a constant pulse-width such that amplitude and frequency were the FES parameters of interest. The model simulated the behaviour of a physiological model which considered the ankle and knee joints, as well as the hamstrings, quadriceps, soleus, gastrocnemius and tibialis anterior muscles during the swing phase of gait. Since FES is the preferred treatment for those affected by foot drop, the primary motivation of the project was to address the device's shortcomings in relation to muscle fatigue and ankle mobility.

2 METHODS

2.1 MODELLING AND SIMULATION

2.1.1 MODEL EQUATIONS AND STATE VARIABLES

The model created considers the tibialis anterior, soleus, gastrocnemius, quadriceps and hamstring. The tibialis anterior is responsible for the ankle dorsiflexion moment. The soleus and gastrocnemius contribute to the ankle plantarflexion moment. Dorsiflexion decreases the ankle angle (the ankle between the foot and shank), while plantarflexion increases the ankle angle. The quadriceps, were included because they are responsible for lifting the entire shank and foot during the early part of the swing phase of gait. The hamstrings were included to make the full model more realistic because they oppose the torque caused by the quadriceps during gait. The upper leg is fixed, while the foot is free to rotate around the ankle and the shank is free to rotate at the knee as both the knee and ankle joint can be treated as pin joints. This can be seen clearly in Figure 1.

The states of the this system can be seen in Equation 1. x_1 is the angle of the ankle. x_2 is the angular velocity of the ankle. x_3 is the angle of the knee. x_4 is the angular velocity of the knee. x_5, x_6, x_7, and x_8 are the normalized length of the contractile element (CE) of the soleus, gastrocnemius, quadriceps, and hamstrings, respectively. The soleus, gastrocnemius, quadriceps, and hamstrings muscles were all modeled following the Hill-Type muscle model outlined in Millard et al. (2013).

$$x_1 = \theta, \;\; x_2 = \dot{\theta}, \;\; x_3 = \phi, \;\; x_4 = \dot{\phi}, \;\; x_5 = \tilde{l}_S^M, \;\; x_6 = \tilde{l}_G^M, \;\; x_7 = \tilde{l}_Q^M, \;\; x_8 = \tilde{l}_H^M \qquad (1)$$

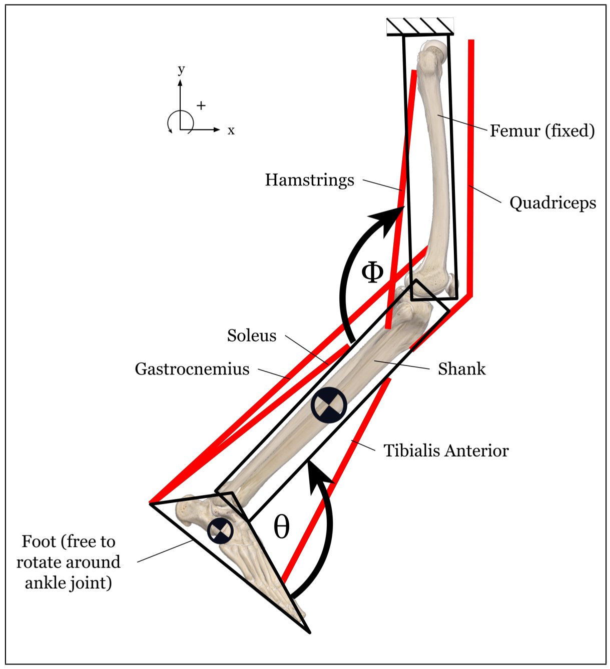


Figure 1: Simplified model of the leg, shank, and foot including the quadriceps, hamstrings, gastrocnemius, soleus, and tibialis anterior muscles used to simulate the effects of varying FES parameters on foot height and muscle fatigue [Image source: SS, 2022].

The corresponding state equations can be seen in Equation 2 below. $\dot{x_1}$ is equivalent to x_2, the angular velocity of the ankle which is the derivative of the ankle angle. $\dot{x_2}$ is the sum of the torques generated by the soleus τ_S, gastrocnemius τ_{GC_a}, tibialis anterior τ_{TA}, and the gravity moment of the foot, all divided by the inertia about the ankle I_{ankle}. The gravity moment of the foot is the product of the distance of the ankle joint to the center of mass d_{COM_a}, the mass of the foot m_{foot}, acceleration due to gravity g, and $\cos(\frac{\pi}{2} - x_3 + x_1)$. $\dot{x_3}$ is equivalent to x_4, the angular velocity of the knee which is the derivative of the knee angle. $\dot{x_4}$ is the sum of the torques generated by the quadriceps τ_Q, gastrocnemius τ_{GC_k} and the hamstrings τ_H and the the gravity moment caused by the shank and foot. The gravity moment of the shank and foot is the product of the product of the distance of the knee from the center of mass d_{COM_a} of the combined segments, the mass of the shank and foot $m_{shank+foot}$, acceleration due to gravity g, and $\cos(\frac{\pi}{2} - x_3)$, all divided by the inertia about the knee I_{knee}. The fifth state equation, $\dot{x_5}$ is the normalized CE velocity of the soleus $\tilde{v}_S^M$ which is a function of the activation of the soleus a_S, the normalized CE length x_5, and the normalized length of the series elastic (SE) element of the soleus, where the SE element is a function of the ankle angle x_1 and the normalized length of the CE element x_5. $\dot{x_6}$, $\dot{x_7}$, and $\dot{x_8}$ are the normalized CE velocities of the gastrocnemius, quadriceps, and hamstrings, respectively. These velocities were derived the same as $\dot{x_5}$ above.

$$
\begin{aligned}
\dot{x_1} &= x_2 \\
\dot{x_2} &= \frac{\tau_S + \tau_{GCa} - \tau_{TA} + d_{COMa} m_{foot} g \cos(\frac{\pi}{2} - x_3 + x_1)}{I_{ankle}} \\
\dot{x_3} &= x_4 \\
\dot{x_4} &= \frac{\tau_Q - \tau_{GCk} - \tau_H + d_{COMk} m_{shank+foot} g \cos(\frac{\pi}{2} - x_3)}{I_{knee}} \\
\dot{x_5} &= \tilde{v}_S^M(a_S, x_5, \tilde{l}_S^T(x_1, x_5)) \\
\dot{x_6} &= \tilde{v}_G^M(a_G, x_6, \tilde{l}_G^T(x_1, x_6)) \\
\dot{x_7} &= \tilde{v}_Q^M(a_Q, x_7, \tilde{l}_Q^T(x_1, x_7)) \\
\dot{x_8} &= \tilde{v}_H^M(a_H, x_8, \tilde{l}_H^T(x_1, x_8))
\end{aligned}
\tag{2}
$$

The torques of the soleus, gastrocnemius, quadriceps, and hamstrings are shown generally in Equation 3 and were determined using the force-length curve of the contractile element and the muscle-tendon equilibrium equations from Millard et al. (2013).

$$\tau = f^M f^T(\tilde{l}^T) d \tag{3}$$

The torque of the tibialis anterior in Equation 4 was calculated as a function of normalized RMS in Equation 5. RMS was calculated using a constant pulse width and the parameters of interest: amplitude and frequency.

$$\tau_{TA} = \beta \widetilde{RMS} f_0^M d_{TA} \tag{4}$$

$$\widetilde{RMS} = \frac{amplitude * \sqrt{frequency * pulse\ width}}{RMS_{max}} \tag{5}$$

2.1.2 MODEL VALIDATION

The constructed Hill-Type muscle model including the soleus, gastrocnemius, quadriceps, and hamstrings muscles was validated through comparison with the Hill-Type muscle model from Millard et al. (2013). The optimal FES parameters used to stimulate the tibialis anterior, including pulse width, amplitude, and frequency, were validated by comparing them with values found in literature. The FES parameter values used in the simulations aligned with the FES parameters investigated in the study by Sabut et al. (2010). The model was also validated by plotting the knee and ankle angles, the heights of both the back and front of the foot, and the normalized muscle lengths of the soleus, quadriceps, gastrocnemius, and hamstrings over the course of the simulation. All three of these graphs underwent a sanity check to verify that certain elements of the model behaved as expected during simulation.

2.1.3 SIMULATION APPROACH

Anatomical values for the mass of the foot and combined mass of the shank and foot were found in literature to be 1.05 kg and 4.35 kg, respectively (Valmassy, 1996). The resting length of the gastrocnemius, hamstrings, quadriceps, and soleus were chosen to be 0.4005 m, 0.4554 m, 0.4554 m, and 0.3501 m, respectively. These parameter values were taken from anthropocentric tables found in Valmassy (1996) assuming a 75 kg subject. The initial parameter values for the ankle and knee angles were chosen to 130 degrees and 150 degrees respectively. Moreover, the simulation started from the toe off phase of gait and lasted 0.39 seconds (Murray et al., 1964). This swing phase duration was determined by Murray et al. (1964) to be a healthy swing phase duration for a healthy male subject. In the MATLAB simulations, ode45 was used to simulate the model behaviour.

2.2 OPTIMIZATION APPROACH

The minimum foot height required to clear the ground was plotted to find the minimum FES parameters that would prevent foot drop. The simulations were executed with a constant pulse width to find the amplitude and frequency that would best minimize muscle fatigue while still achieving the required foot height. The optimized parameters were selected through running 900 simulations with different amplitude and frequency combinations. The 900 combinations were determined by testing 30 different amplitudes evenly spaced between 0 - 120 V each with 30 different frequencies evenly spaced between 10 - 50 Hz. A fitted curve of the minimum possible amplitude and frequency combinations that would maintain a foot height above 0 meters was determined from the 900 simulations of the swing phase. All of the amplitude and frequency combinations along this curve were then used to calculate the muscle fatigue over the course of the swing phase caused by the FES. The combination that caused the least muscle fatigue while still ensuring the foot would not drop below 0 metres in height was selected as the ideal parameter values.

One of the major limitations of FES is that stimulated muscles tend to fatigue very rapidly, which limits the role of FES in applications such as standing and walking (Thrasher et al., 2005). Muscle fatigue can be measured by looking at the normalized fatigue time integral, FTI, as seen Equation 6, where T is time for the trial, F^M is the force of the muscle and ${f_o}^M$ is the maximum isometric force of the muscle. Previous studies concluded that a low amplitude voltage and low frequency reduces the muscle fatigue time Kumar et al. (2008). Therefore, the smallest amplitude and frequency combination that adequately lifts the foot off the ground was considered optimal for minimizing muscle fatigue.

$$FTI = \frac{\int_0^T (F^M) dt}{f_0^M} \tag{6}$$

2.3 APPROACH TO ANALYSIS

Since both the knee and ankle angles are states of the system, x_1 and x_3 were plotted over time to visualize how each segment of the model moved with respect to each other throughout the course of the simulation. Next, heights of both the front and back of the foot at a specific time instance were determined using Equation 7 and Equation 8, respectively. These values were then plotted over time to visualize the height of the foot during the simulation. The minimum height of either the front or back of the foot was also recorded for each simulation and were used in the optimization of the frequency and amplitude parameters. Additionally, since the normalized muscle lengths of the soleus, gastrocnemius, quadriceps, and hamstrings were all states of the system, x_5, x_6, x_7, and x_8 were all plotted over time to visualize how each segment of the model moved during the simulation.

$$Foot\ Height_{back} = Knee\ Height - (0.4\cos(\pi - x_3) + 0.2\sin(\frac{\pi}{2} - x_3 + x_1)) \tag{7}$$

$$Foot\ Height_{front} = Knee\ Height - (0.4\cos(\pi - x_3) - 0.05\sin(\frac{\pi}{2} - x_3 + x_1)) \tag{8}$$

To analyze the validation plots of the simulation, trends observed in the plots were used to compare how each segment of the model moved during the swing phase to what was expected in Valmassy (1996).

3 RESULTS & DISCUSSION

3.1 OPTIMIZATION

The swing phase simulation was run for 900 evenly spaced frequency and amplitude combinations and then the minimum foot height (front or back) found over the course of every simulation was plotted in Figure 2. The blue, green, and orange sections of the graph show amplitude and frequency combinations which were inadequate since the minimum foot height during those simulations did not remain above 0 meters (the ground). The yellow plane indicates a minimum foot height of 0 meters about the ground for all combinations of amplitude and frequency. The sections of the graph that intersect the yellow plane indicate amplitude and frequency combinations which maintained a minimum foot height of 0 meters above the ground. Figure 3 shows the bottom view of the graph in Figure 2 to better visualize the amplitude and frequency combinations which achieved a minimum foot height of 0 meters above the ground.

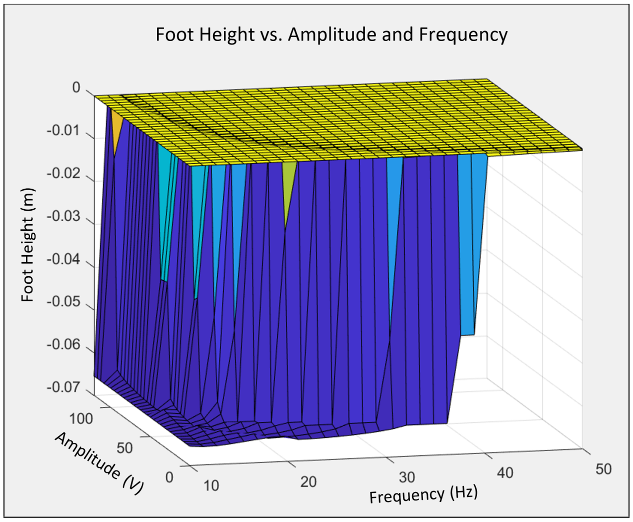


Figure 2: Minimum foot height above the ground during 900 simulations of the swing phase with 900 different evenly space amplitude and frequency combinations [Image source: SD, 2022].

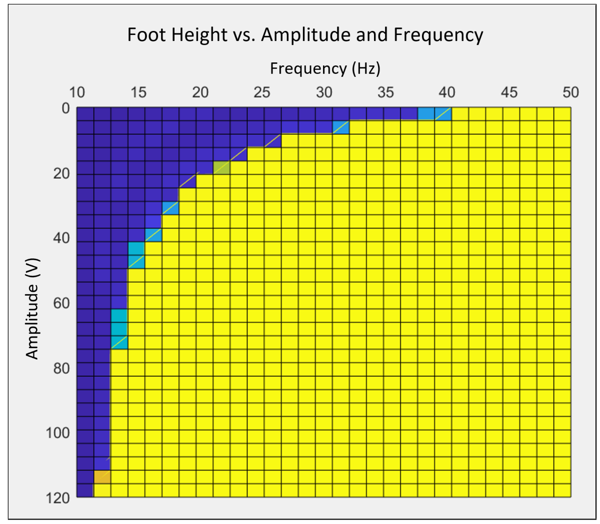


Figure 3: Bottom view of the 3D foot height vs. amplitude and frequency graph to better visualize the relationship [Image source: SD, 2022].

Figure 4 shows a scatter plot of amplitude and frequency combinations that achieved a minimum foot height of 0 meters above the ground and bordered the combinations which achieved lower minimum foot heights. A least squares curve fitting method was used to determine the exponential line of best fit for the amplitude and frequency combinations. The line of best fit was determined to be $y = 345.67e^{-0.119x}$, where y is amplitude and x is frequency. This relationship between amplitude and frequency was plotted in the purple curve in Figure 5. Muscle fatigue calculations were performed using the amplitude and frequency combinations along this line of best fit to visualize the relationship between fatigue, amplitude and frequency in the blue curve of Figure 5.

The amplitude and frequency combination that produced the minimized muscle fatigue while maintaining a minimum foot height of 0 metres was selected as the optimal FES parameters. The selected amplitude and frequency were 2.177 V and 50 Hz, respectively. By selecting amplitude and frequency values that minimize foot height of the swing phase above 0 meters, foot drop is treated while minimizing the risk of muscle fatigue caused by overstimulation (Kumar et al., 2008). These parameter values also promote a more natural gait pattern, as they provide the least amount of external activation on the patient by FES required for the foot to clear the ground.

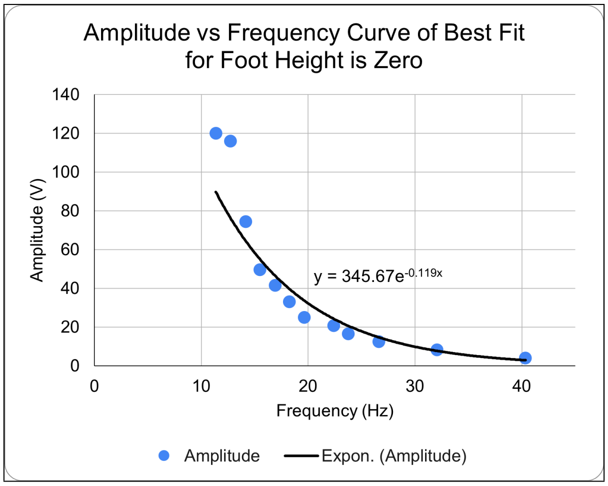


Figure 4: Scatter plot of the boundary amplitude and frequency combinations that resulted in a minimum foot height of 0 meters with a curve of best fit in black on top [Image source: SD, 2022].

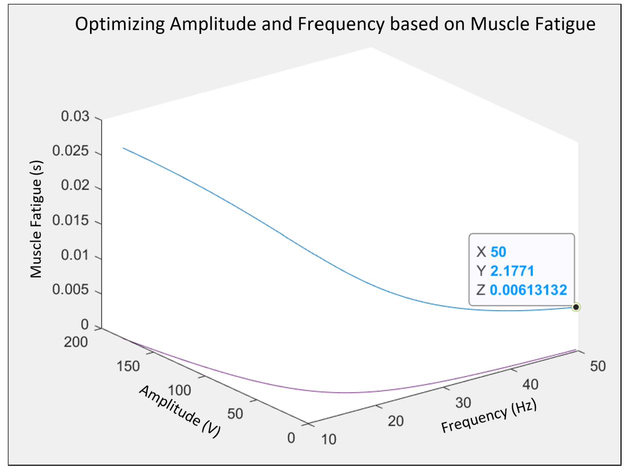


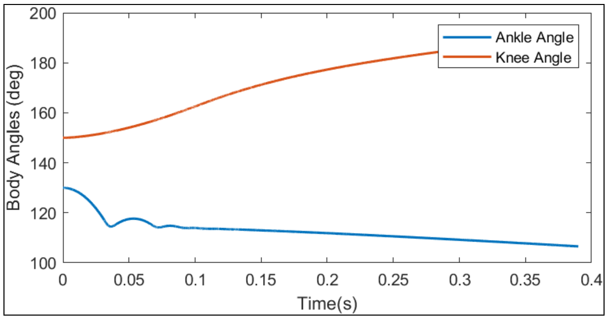
Figure 5: Muscle fatigue plotted as a function of the amplitude and frequency combinations (blue) along the curve of best fit (purple) [Image source: SD, 2022].

3.2 VALIDATION

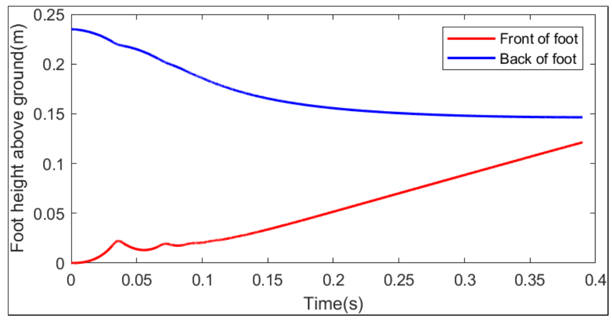
The swing phase simulation was run with an FES amplitude of 2.177 V and frequency of 50 Hz. Figure 6 shows a collection of the resulting relationships plotted to validate the model and simulation. Figure 6a shows how the knee and ankle angles change over the course of the simulation. Initially, the angle of the ankle joint relative to the shank is decreased and gradually increases, which can be attributed to the dorsiflexion of the ankle and relaxation as the foot approaches the ground. When considering the knee angle it increases throughout the swing phase which is associated with flexion at the beginning of the swing and extension at the end of the swing.

Figure 6b shows the height of the back and front of the foot from the ground over the course of the simulation. The height of the front of the foot increases upwards from the ground over time which is attributed to the lifting of the front of the foot during dorsiflexion of the ankle. Dorsiflexion of the ankle is caused by the activation of the tibialis anterior muscle by the FES signal. The height of the back of the foot decreases slightly before stabilizing around 0.15 m above the ground during the swing phase of gait. The initial decrease is a result of the dorsiflexion at the ankle, which causes the heel of the foot to move downwards. The foot height results would have been different if the thigh were not immobilized in the model. In reality, the thigh swings forward during the swing phase of gait which contributes to a greater decrease in foot height of the back of the foot and a greater increase in foot height of the front of the foot.

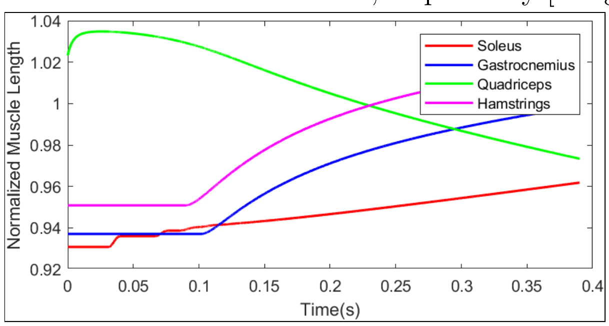
Figure 6c shows the normalized muscle lengths of the soleus, gastrocnemius, quadriceps, and hamstrings muscles over the swing phase simulation. Over the duration of the swing phase, the normalized muscle lengths of the soleus and gastrocnemius increased. These muscles were under tension as a result of the shortening of the tibialis anterior. The normalized length of the hamstrings increased over the duration of the swing phase, as they are under tension and oppose the extension at the knee. Moreover, the normalized length of the quadriceps decreased over the duration of the swing phase, as their activation produced extension at the knee. Overall, the behaviour of the model as a result of FES stimulation during the swing phase aligns with what is expected during natural gait. The plots served as validation that the selected FES parameters addressed the clinical problem; foot drop.



(a) Knee and ankle angle in degrees as functions of time plotted in red and blue, respectively [Image source: SD, 2022].



(b) Heights of the back and front of the foot in meters as functions of time plotted in blue and red, respectively [Image source: SD, 2022].



(c) Normalized muscle length of the soleus, gastrocnemius, quadriceps, and hamstrings muscles as functions of time [Image source: SD, 2022].

Figure 6: Results from the validation simulation of the swing phase using the optimal FES amplitude and frequency values.

4 CONCLUSION & RECOMMENDATIONS

Through completing the planned simulations, optimal FES parameters corresponding to a minimized muscle fatigue were selected. The chosen frequency, amplitude, and pulse width were successful in combating foot drop which was verified by interpreting the plotted results. When considering the initial objective of determining the optimal FES parameters that minimize muscle fatigue while still combating foot drop, the project was successful.

For future iterations of this project, several factors would be further explored and implemented to enhance the simulations. Due to the time constraint, our simulations focused on selecting ideal values for the frequency and amplitude of the FES signal and disregarded pulse width. Studying pulse width's relationship to dorsiflexion of the tibialis anterior and the lifting of the foot off the ground was omitted since both pulse width and amplitude are involved in muscle recruitment and were expected to elicit the same result (Ferrarin and Pedotti, 2000), (Jeon and Griffin, 2018). Thus, in future iterations, it would be beneficial to study the relationship of pulse width with foot height in conjunction with the other two parameters to verify that the initial assumption was valid. Pulse width is also directly related to how the FES signal is delivered to the muscle, either by a step function or a ramp function (Ferrarin and Pedotti, 2000). Due to the time limitation on the project, the simulations only considered step functions, in future iterations, the simulations should examine the effects of ramp-up and ramp-down functions and their effect on muscle fatigue and foot height. The full-model used in the simulations which included five muscles, the ankle and the knee joint would be classified as a simplification of the physiological components involved in gait. The full-model fails to address the role of the hip joint and hip abductors which allow the movement of the entire leg, including the upper leg during the swing phase which is considered fixed in the full model. In addition to including the hip joint and hip abductor muscles to the model, the full gait cycle should also be considered. Considering the whole gait cycle would allow for the simulations to more closely mimic natural gait as the behaviour of the muscles during the transitions from the stance phase to the swing phase and again to the stance phase would be able to be examined. This could be extended to simulating multiple gait cycles. Conducting further research regarding muscle fatigue due to cyclic loading by FES would allow for further examination of how the optimized FES parameters compare to those used in literature. It would also allow for determining the useful duration of the FES device before a certain muscle fatigue threshold is reached. Determining the useful duration threshold would allow for quantifiable results that would decipher whether the optimized FES device outperforms currently available devices.

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