Drop Foot and Functional Electrical Stimulation: An Integrated Model

Daniel De Sousa - 20608147
Sam Feng - 20604727
Robyn Klassen - 20622697
Vincent Shadbolt - 20617236
Sunny Willert - 20619549

Faculty of Engineering
Department of Systems Design Engineering

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1 Abstract

Many knee and tibialis anterior (TA) surgery patients, as well as those afflicted by progressive and non-progressive nervous disorders such as multiple sclerosis and strokes, are affected by a condition referred to as drop foot. The condition is characterized by reduced or total inhibition of the nervous activation signals sent to dorsiflexor muscles, such as the TA, inhibiting proper upward movement of the foot during the swing phase of the gait cycle. One form of treatment, functional electrical stimulation (FES), provides dynamic artificial electrical signals to the muscle and its motor nerves to induce muscle contraction [1]. As such, the development of this treatment requires an understanding of the technique's effects on the foot. To achieve this goal, the following paper delves into the development and integration of two models. The first characterizes an optimal FES muscle activation signal that simultaneously minimizes induced muscle fatigue. The other characterizes the biomechanical movement of the foot as a function of this signal, and is fundamentally based on both the behaviour of a double-pendulum and the Hill-type model. Mathematical modelling of the integrated swing phase model using MATLAB's ode45 function permitted visualization of the optimal FES operation. The model was validated dynamically by calculations of the generated muscle force, toe elevation from the ground, and trajectories of the COM of the foot through space. In this way, it was found that the optimized FES activation signal, operating at 34Hz, was sufficient in lifting and holding the foot above the ground, alleviating the patient experience of drop foot.

2 Introduction

Drop foot is a condition referring to weakened dorsiflexion of the toe and ankle, as a consequence of reduced or inhibited contraction of the muscles controlling the movement. Individuals affected by drop foot have difficulty lifting the distal end of their foot, and so must often compensate by exaggerating knee and hip flexion while they walk; this prevents their toes dragging along the ground during the swing phase of the walk cycle. Muscles responsible for dorsiflexion include the TA, extensor hallucis longus, and extensor digitorum longus. They ensure clearance of the foot above the ground when a person takes a step, and stabilize plantar flexion of the foot whenever the heel strikes the ground. Drop foot may be attributed to various factors, including strokes, damage to the muscles involved in dorsiflexion, bone, cyst, or tumour overgrowths, neuropathies, and diabetes. These factors may be classified into three groups: neurological, resulting from damage to the brain or spine; muscular, caused by diseases including polio or muscle dystrophy; anatomic, due to nerve injuries to the spine or leg [1].

The preliminary objective of this project is to determine and develop a mathematical model capturing the biomechanics of normal walking and of drop foot. This will fuel the primary objective, which is to determine the optimal FES signal to the TA that will generate minimal muscle fatigue and evaluate whether or not the signal is sufficient for mitigating drop foot. In achievement of these two objectives, an understanding of the overall mechanics of the foot and anatomy involved in the human gait cycle can be obtained.

The biomechanics of the model developed in this report is heavily based on the double pendulum, which helps to depict the swing phase of the gait cycle, as demonstrated by E. Have et al. in “The Double Pendulum” [2]. In the model developed for this study, the behaviour of the shank and foot was represented by a double pendulum, with a fixed knee joint. Additionally, the model also employs the damped equilibrium version of the Hill-type muscle model as defined by Delp et al. in “Flexing Computation Muscle: Modelling and Simulation of Musculotendon Dynamics” [3]. Composed of a contractile element (CE), a series element (SE), and a parallel element (PE), the Hill-type muscle model can predict and determine the behaviour of the muscles of interest when under contraction. Furthermore, the solution developed as part of this report is based on TA activation data during the entire walk cycle obtained from Barela et al. [4].

The context of this report is of great importance as many people are afflicted by this condition, and as such, it is a prevalent issue that should be addressed. Drop foot affects up to 13% of patients that have received a surgery involving the TA, as well as up to 4% of patients that undergo surgeries to the knee, for example, total knee arthroplasty [5]. Moreover, another motivation for this project regards the group’s interest in this problem space and developing solutions to medically-oriented problems.

3 Methods

To form an understanding of FES signal design and to assist people with drop foot during the swing phase, two models were developed. The first was a model of TA stimulus during the swing phase of the gait cycle and the second was a biomechanical model of the shank, ankle and foot based on double pendulum dynamics. These models were then merged to construct an accurate drop foot simulation.

3.1 Activation Model

Variation in ankle angle is controlled by the antagonistic muscles TA and soleus which cause dorsiflexion and plantar flexion of the foot respectively. As the lack of dorsiflexion is the principal issue in drop foot, the activation of the TA is assumed to be zero for the purpose of this study. Thus, to correct for drop foot during gait, a model of the activation of the TA in a healthy subject is required. It is understood that the soleus is activated minimally during the swing phase, its primary purpose being to stabilize the movement of the ankle. As such, the development of the muscle activation model focuses on the stimulation of the TA. **Figure 1** displays the general EMG trend that the TA exhibits during the full gait cycle, as found by Barela et al., which will form the basis on which muscle activation is modelled [4]. Note that the solid black line characterizes the raw walking EMG data of interest, having undergone no additional analysis or data manipulation.

Notably, the independent variable of the graphs depicting EMG data is the percentage of walking cycle, which is converted into time (seconds) by assuming a predetermined gait cycle time of 1.41s [6]. In doing so, an activation curve for the TA was modelled by performing a rolling regression on each obtained sample time. For each point, an array of independent and dependent variables were obtained using past and future data of the current time. A quadratic regression was then run on the resulting data interval. The regression model for TA is depicted in **Figure 2**, with the red lines denoting the regression lines.

Validation of the activation model has two stages: sanity checks and variation checks. The former involves visual assessment of the figures below to classify the degree of the polynomial - for example, linear, quadratic or cubic - throughout the gait cycle. This allows for a general understanding of the variation of regression fit required, providing the most accurate polynomial degree to be identified. The latter check involves identifying the coefficient of determination (R-squared) of each regression line (for each degree of polynomial), yielding a quantitative description of the deviation between the regression fit and the sample data, in addition to the amount of variance the regression accounted for. Both validation stages established that linear regressions underfit the data, as the coefficients demonstrated an inability to curve with the sample data, while the cubic regression overfit the data. Thus, it was determined that the quadratic regression established the most accurate activation model.

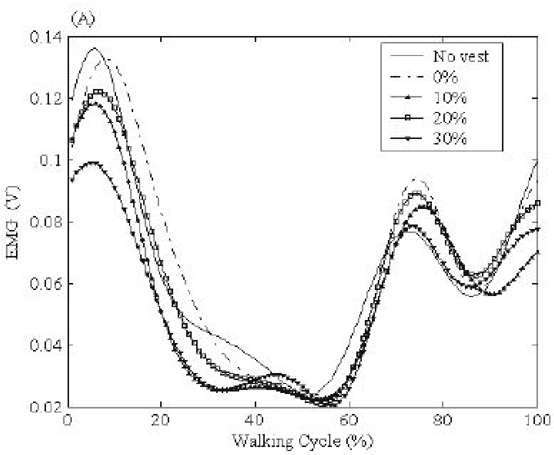


Figure 1. Mean trajectories of EMG from TA during the walking cycle [4].

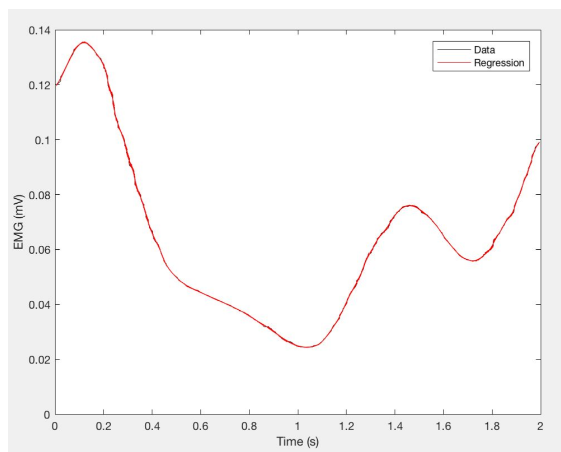


Figure 2. Regression modelled trajectories of EMG from TA during the walking cycle.

Then, by utilizing a set of simulated data from the initial regression model, an FES model was developed. Given that the nature of muscle activation is characterized by neurological signals with oscillating peaks and troughs of varying amplitudes, the model was aimed at recreating this behavior. Note that the peaks of this model were aligned with that of the regression modelled activation. The parameters taken into consideration when developing the FES peaks include the frequency, the pulse width, duty cycle, and amplitude of the stimulus.

The frequency of the stimulation signal defines the number of stimuli peaks occurring in a second. In a study performed by B. M. Doucet et al., it was stated that "electrical stimulation is typically provided at higher frequencies (20-50Hz) expressly to produce muscle tetany and contraction that can be used for functional purposes" [7]. In the same study, it is established that a higher frequency stimulus with lower relative amplitude is found to be the most comfortable stimuli. This ensures that the stimulus is less noticeable during continuous application and thus more tolerable for the patient, as increases in frequency and amplitude result in more forceful activation of the muscle [7]. Pulse width, another signal factor to consider, is the time duration of the stimulus. In a study performed by D. F. Collins, it was determined that a short narrow pulse duration between 500 and 1000 microseconds would result in a less fatigued muscle, whereas wide pulses stimulated larger contractions with a greater central distribution [8]. Another parameter of activation is the duty cycle of a stimulus, defined as the percentage or ratio of time of an activated stimulus throughout its pulse width [7]. Typically, "common clinical applications use a 1:3 duty cycle as a standard" [7]. AG Bracciano agrees with this, stating that the optimal duty cycle to reduce fatigue in a patient's muscle is to use a stimulus with a duty cycle of approximately 25% [9]. The final parameter to consider is the amplitude or intensity of the stimulus. As the amplitude increases, the strength of the depolarizing effect found within the muscle also increases. A higher percentage of muscle fibres activate, thereby generating a stronger contraction [10]. A consistent optimal amplitude of a stimulus does not need to be established, as the intensity varies greatly between different muscles and during different portions of the gait cycle; therefore, the amplitude of the applied TA stimulus is required to vary during the gait cycle.

However, to develop a model of an FES signal that could be applied to the TA during a gait cycle, some assumptions were made regarding the stimulus parameters. Specifically, it was presumed that the frequency of the stimulus would be equal to the inverse of the pulse width of the stimulus, resulting in the stimulus being constantly applied. Furthemore, since FES and natural muscle stimulation consist of various sharp peaks and troughs, rather than a typical uniform step function, the duty cycle of the stimulus was considered irrelevant. Therefore, the frequency and amplitude of the stimulus signal were the only two input parameters of the FES model.

To model the activation of muscles during the gait cycle, the location of signal peaks and troughs needed to be identified based on the frequency. Peaks are located a pulse width apart, as are troughs, since pulse width is equivalent to the inverse of frequency. Each adjacent peak and trough were regressed to find their linear relation. All time data points that fell between a respective peak and trough had their activations calculated using the associated linear relation. **Figures 3a** and **3b** demonstrate the basic shape in which the stimulus is formed.

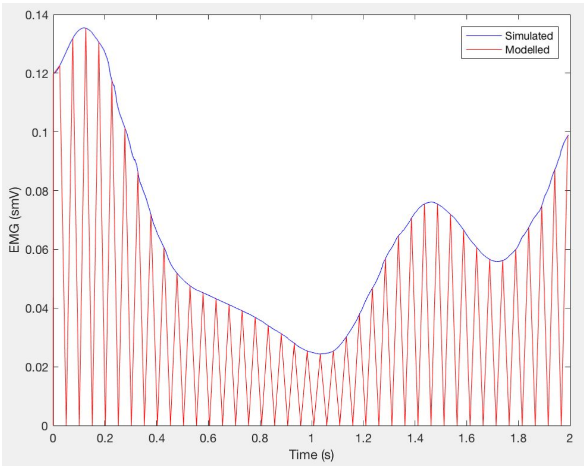


Figure 3a. FES signal model of walking gait cycle at 20Hz.

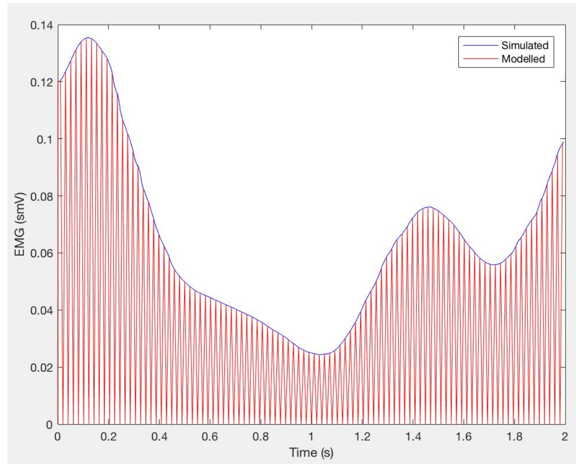


Figure 3b. FES signal model of walking gait cycle at 50Hz.

As the regression model defines the optimal stimulus amplitude throughout the gait cycle, fully defining the activation model requires identifying the frequency at which the relative muscle fatigue is minimized. As stated previously, FES is most effective when stimulus frequency is within a 20-50Hz range as that ensures minimal fatigue induced.

Fatigue exhibited by a muscle is calculated by determining the absolute integral under its force curve [11]. Although force data for the TA during the gait cycle was not available for this study, “Mechanical factors affecting the estimation of TA force using an EMG-driven modelling approach” by Stuart Miller, noted that there is a “curvilinear relationship between activation (surface EMG) and muscle force” [12]. Therefore, according to this relationship, the relative fatigue between stimuli at different frequencies can be characterized using the absolute integral of the activation-time curve [12]. Then, to find the optimal frequency, all integer values within the range of 20Hz to 50Hz were individually run through the FES signal model to find the relative fatigue. **Figures 3a** and **3b** above display the end points of this frequency range at 20Hz and 50Hz respectively and **Figure 4** shows the relationship between the relative fatigue and the frequency of the stimulus. **Figure 4** also demonstrates that the relationship between fatigue and frequency is erratic, with sparse optimal frequencies throughout the range. Despite this, the optimal frequency at which the FES signal should occur was determined to be 34Hz, as it decreases the relative fatigue exerted on the TA. However, it is noted that the change in relative fatigue between the different frequencies is not very large, and there are other minimal frequency values that would also minimize fatigue. This suggests that, if the optimal value selected does not fit the biomechanical model well, other frequencies are available for testing.

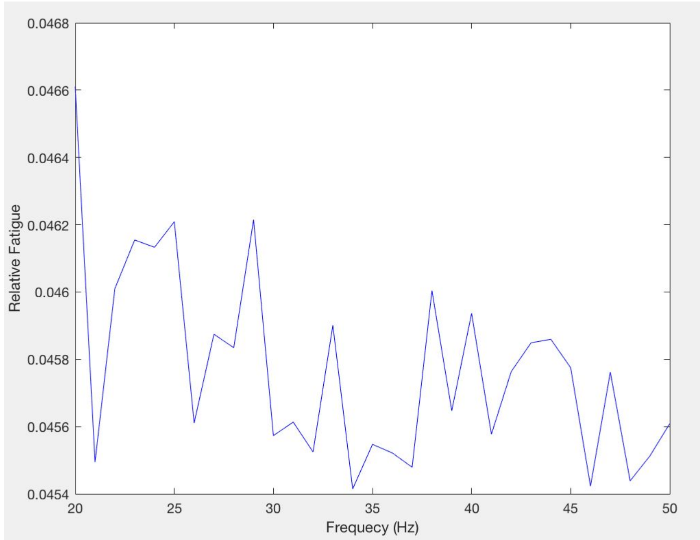


Figure 4. Relative fatigue for varying stimulus frequencies.

It is worth noting that fundamental differences exist between FES signals and the natural stimulation of the muscle. Physiologically, when the nervous system stimulates a muscle, the motor units contained within the muscle are activated asynchronously [13]. This means that the motor units are rotated through during stimulation, and the number which are stimulated is dependent on the magnitude of activation [13]. Conversely, FES signals recruit motor units synchronously, meaning all are stimulated simultaneously [13]. Therefore, to properly contract the muscle, a higher frequency is required than in natural stimulation, leading to increased fatigue rates for two reasons [13]. First off, increasing the frequency of stimulation results in motor units being activated more often, depleting stored energy (ATP) within the muscle at a greater rate. Secondly, natural asynchronous motor unit activation allows for the opportunity of fatiguing motor units to be replaced by those currently not in use. However, FES signals stimulate the motor units synchronously, resulting in simultaneous activation, preventing this replacement ability and thereby increasing the rate of fatigue. Therefore, it is necessary to ensure that the optimal frequency is great enough to produce tetanic contractions, while not being so high as to over-stimulate the muscle and increase the rate of fatigue [13].

Upon implementing the activation model into the biomechanical model of the shank and foot, the activations obtained from the FES signal model are clipped to select only the swing phase present at 60-100% of the walking cycle, as shown in **Figure 5**.

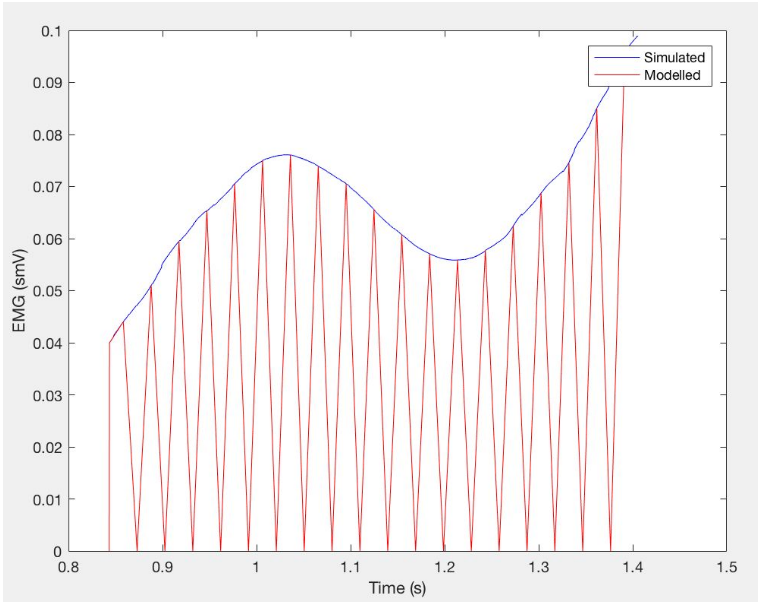


Figure 5. FES signal model for swing phase at optimal frequency of 34Hz.

A comparable literature source of control systems for FES is "Functional electrical stimulation" provided by Lynch and Popovic. The paper describes stimulation as having the shape of a pulse train, as seen below in **Figure 6**. Here, the typical waveform is demonstrated as having an amplitude A, pulse widths C and D (with D being present in some cases, it is not considered here) and frequency f_{stim}. There also exists a duty cycle which is represented as the percentage of period utilized as the stimulation pulse [13]. Contrary to this paper however, it was decided that a triangular function would be of greater accuracy to healthy muscle stimulation than a pulse train. This decision is supported by EMG activation which exemplifies a spiking behaviour, described as frequency peaks and troughs, during muscle stimulation. Thus, a triangular function with varying amplitude would be best suited for this model, since it better reflects true EMG behaviour.

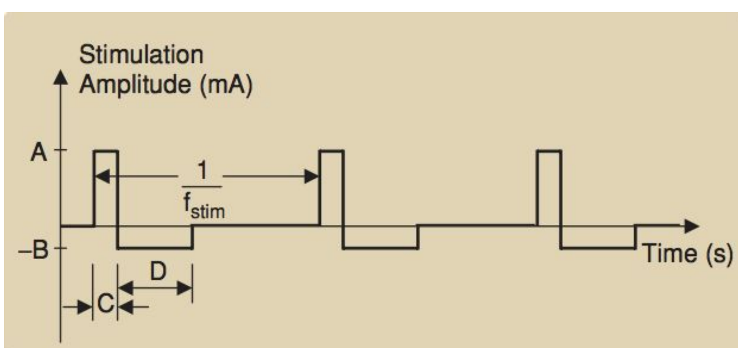


Figure 6. Stimulation pulse train [13].

3.2 Biomechanical Swing Phase Model

The behaviour of the double pendulum is frequently used when modelling the biomechanics of the swing phase of gait cycle [2]. The two pendulums in question, however, typically represent the thigh and the shank. In this case, for the present development of the swing phase model, the pendulums will instead predict the behaviour of the shank and the foot [14]. Consequently, the fixed pivot point of the pendular model is the knee joint rather than the hip joint, when compared to these similar models. As an additional factor to consider, the second pendulum - in most models - is commonly represented as a simple rod with mass whereas with the implemented swing phase model, the second pendulum represents the foot, which is approximated by a triangular shape. In this part of the model, Lagrangian mechanics help in defining the predicted motion of the body segments.

The developed swing phase model also utilizes the damped equilibrium version of the Hill-type muscle model to understand the behaviour of muscle contraction [3]. Composed of a CE, SE, and PE, the behaviour of muscles under contraction can be determined and predicted dynamically. In the developing swing phase model, this is particularly relevant as the torque provided by the muscle contraction of the TA affects the model dynamics of the shank and foot. To establish the CE force-length and force-velocity curves, data was collected from Lieber et al. and Delp et al. respectively [15] [3] and regressions were performed on this data.

The state variables utilized by the biomechanical model are described by **Equations 1** through **6**, including both the angle and angular velocity of the shank about the knee, the angle and angular velocity of the foot about the ankle, and CE lengths of the TA muscle. **Equations 7** and **8** depict the Lagrangian equations that determine the angular accelerations of components in a double pendulum. The state vector, **x**, is defined as $[x_1\ x_2\ x_3\ x_4\ x_5\ x_6]'$, where x_1, x_2, x_3, x_4, x_5 and x_6 represent the angle of the shank from the negative Y-axis, the angle of the foot's center of mass (COM) from the negative Y-axis, the angular velocity of the shank, the angular velocity of the foot's COM, the CE length of the TA, and the CE length of the soleus respectively. In the equations below, I_1 represents the moment of inertia of shank about a perpendicular axis through the knee joint, I_2 represents the moment of inertia of the foot about a perpendicular axis through the ankle joint, and $I_{2,COM}$ represents the moment of inertia of the foot about the COM of the foot. The change in the angle, demonstrated by $\dot{x}_1$ and $\dot{x}_2$, can be represented by x_3 and x_4, and equivalent to the respective angular velocities. The change in angular velocity of the shank, represented by $\dot{x}_3$, is equivalent to the angular acceleration of the upper pendulum component as determined by Lagrangian mechanics (**Equation 7**). The change in angular velocity in the foot's COM, as represented by $\dot{x}_4$, similarly incorporates angular acceleration from Lagrangian mechanics for the lower pendulum component, which is given by **Equation 8**. The total $\dot{x}_4$ (**Equation 4**) is equivalent to the sum of **Equation 7** and the acceleration generated by muscle moments, which is calculated by dividing the moment created by the TA and soleus by the moment of inertia of the foot about the ankle. The change of CE lengths of the TA and soleus, $\dot{x}_5$ and $\dot{x}_6$, are calculated using the Hill-type muscle model, shown in **Equations 5** and **6.**

$$\dot{x_1} = x_3 \tag{1}$$

$$\dot{x_2} = x_4 \tag{2}$$

$$\dot{x_3} = \ddot{\theta}_{LAGRANGIAN,1} \tag{3}$$

$$\dot{x_4} = \ddot{\theta}_{LAGRANGIAN,2} + \frac{M_{TA}}{I_{2,COM}} \tag{4}$$

$$\dot{x_5} = v_{TA}(a_{TA}, x_5, \tilde{l}_{TA}(\pi - x_1 - x_2)) \tag{5}$$

$$\dot{x_6} = v_S(a_S, x_6, \tilde{l}_S(\pi - x_1 - x_2)) \tag{6}$$

$$\ddot{\theta}_{LAGRANGIAN,1} = \frac{\begin{gathered}-[m_2 l_1 l_2 x_4^2 \sin(x_1-x_2) I_2 + m_2^2 l_1 l_2^3 x_4^2 \sin(x_1-x_2) + m_1 g d_1 \sin(x_1) I_2 + m_1 g d_1 \sin(x_1) m_2 l_2^2 \\ + m_2 l_2^2 + m_2 g l_1 \sin(x_1) I_2 + m_2^2 g l_1 \sin(x_1) l_2^2 + m_2^2 l_1^2 l_2^2 \cos(x_1-x_2) x_3^2 \sin(x_1-x_2) \\ - m_2^2 l_1 l_2^2 \cos(x_1-x_2) g \sin(x_2)]\end{gathered}}{-m_2^2 l_1^2 l_2^2 \cos(x_1-x_2)^2 I_1 - I_2 + I_1 m_2 l_2^2 + m_2 l_1^2 I_2 + m_2^2 l_1^2 l_2^2} \tag{7}$$

$$\ddot{\theta}_{LAGRANGIAN,2} = \frac{\begin{gathered}[l_1^2 \cos(x_1-x_2) m_2 l_2 x_4^2 \sin(x_1-x_2) + l_1 \cos(x_1-x_2) m_1 g d_1 \sin(x_1) + l_1^2 \cos(x_1-x_2) m_2 g \sin(x_1) \\ + l_1 x_3^2 \sin(x_1-x_2) I_1 + l_1^3 x_3^2 \sin(x_1-x_2) m_2 - g \sin x_2 I_1 - g \sin(x_2) m_2 l_1^2] m_2 l_2\end{gathered}}{-m_2^2 l_1^2 l_2^2 \cos(x_1-x_2)^2 + I_1 I_2 + I_1 m_2 l_2^2 + m_2 l_1^2 I_2 + m_2^2 l_1^2 l_2^2} \tag{8}$$

Parameters relating to body dimensions may be defined anthropometrically, taking into account several assumptions, as shown in **Figure 7**. According to Bogdan and Costea, the average total length and height of a male right foot is 0.2921m and 0.08798m respectively [16]. The location of the COM of the foot was determined examining ratios from the COM to the foot. Since the foot is approximated by two right triangles, the COM may be located at approximately ⅓ of the total foot height, measured from the ground. Similarly, the COM of the foot may be approximated as being 43.6% of the total foot length from the heel, as given by Bluestein et al.; the COM of the shank as being 44.3% from the knee joint is also approximated by these authors [17]. The horizontal distance from the ankle to the heel was determined from anthropometric measurements on group members, and determined to be 0.0762m. The remainder of the anthropometric values for the foot were solved using trigonometry, and are represented in **Figure 8**. A summary of the anthropometric values used in in this model is given in **Table 1** in the Appendix.

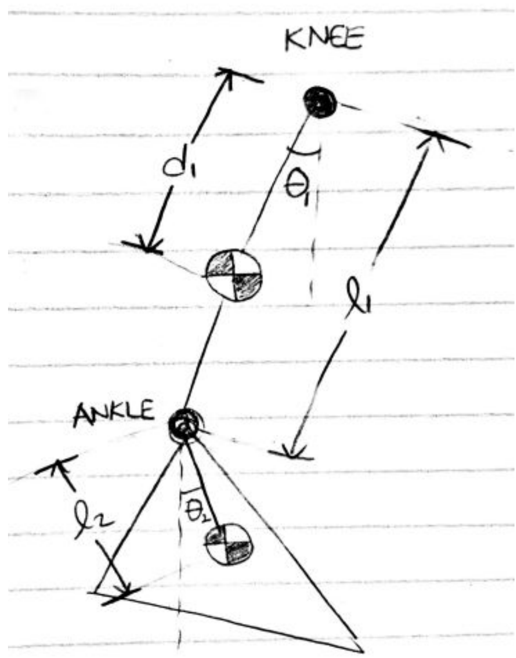


Figure 7. Double pendulum used for biomechanical swing phase model.

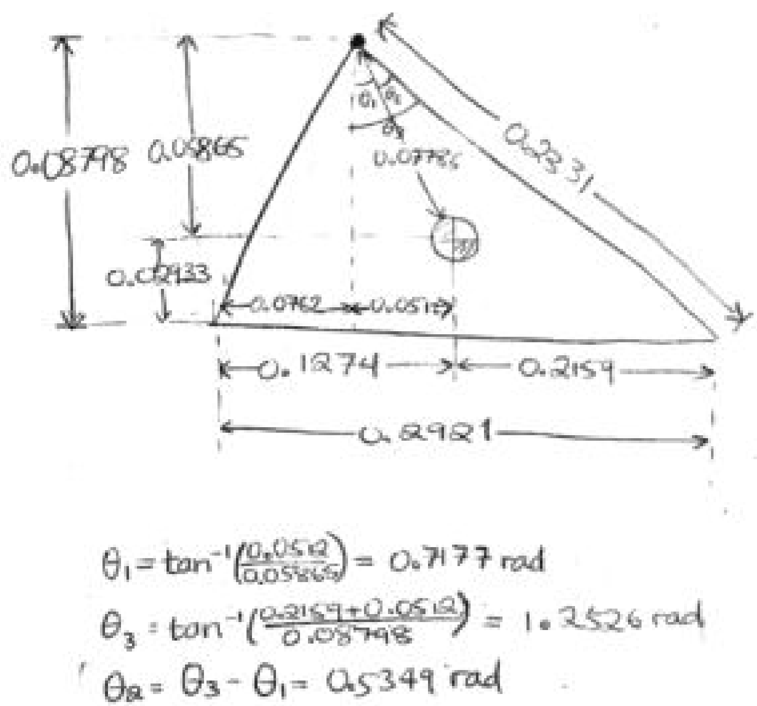


Figure 8. Anthropometric parameters of an average male foot.

Given that the swing phase begins at the 60% mark of the walk cycle, the initial shank angle from the negative Y-axis - fixed at the knee joint - is determined by Bianchi et al. to be approximately 5.926 radians. Similarly, the initial ankle angle from the negative Y-axis - fixed at the ankle joint - is determined to be 6.574 radians [18]. The angular velocity of the shank at the beginning of the swing phase was established by Lau et al. to be 2.042 rad/s and the angular velocity of the ankle was set as 2.460 rad/s [19]. However, for patients suffering from foot drop, it was noted by Błażkiewicz that the angular velocities of both the shank and ankle were 47% that of a standard cycle [20]. Thus, the shank and ankle angular velocities were set to 0.960 rad/s and 1.156 rad/s respectively. Based off of Arnold et al., at the time of swing phase, the normalized CE soleus length is 0.8129m and the normalized CE TA length is 1.045m [21]. Taking into consideration the average mass of North American, 80.7kg, and the average percentages of weight of the foot and the shank in the body, the mass of the foot and shank are 1.11kg and 3.47kg respectively [22] [23]. These values are also summarized in **Table 1**. The values for the maximum isometric force and the TA moment arm are provided by Professor Tripp.

When constructing the biomechanical model, several assumptions were made to feasibly model the swing phase of the gait cycle. To avoid modelling the mechanics of the remainder of the body, the shank and and the foot are considered isolated components, with the knee considered as a fixed joint about which the shank rotates during swing phase. With regards to the foot, anthropometric data not usually provided by studies was required for accurate modelling. As such the foot is approximated by a perfectly triangular shape, to determine these values. For simplicity, when calculating the normalized CE velocity of muscles, the pennation angle and damping coefficient are assumed to be 0 and 0.1 respectively. Another assumption made is that the swing phase, the period between when the toes push off the ground and the heel strike, is the 60-100% portion of the gait cycle. The activation of the soleus muscle is also presumed to have a value of zero throughout the duration of swing phase.

4 Results

After implementing and integrating these two models, two sets of simulations were run. The first simulation is based off a patient afflicted with drop foot, due to a lack of TA activation. The second simulation was based off the same patient with the added complexity of the external stimulation of the TA. The simulations were run for 0.65 seconds as a means to mimic the swing phase. Note that in simulation one, the TA activation was set to zero whereas for simulation two, the TA activation was based on the data points from the activation model.

Figures 9a and **9b** below are screen captures of the animated simulation immediately prior to the end of the swing phase. Here, the end point on the third linked segment marks the location of the patient's toes and the solid horizontal black line represents the ground. In this case, the axes are in reference to the knee.

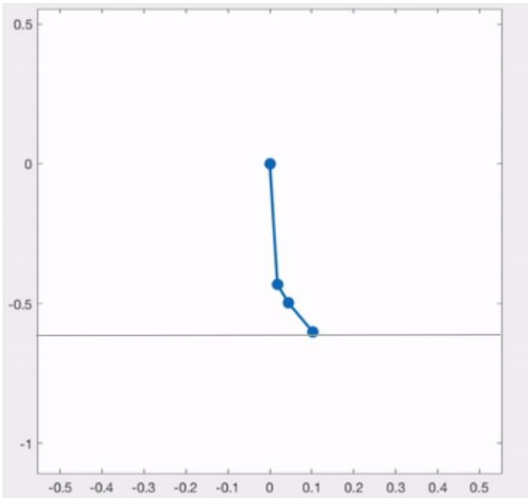


Figure 9a. Foot drop patient simulation without TA stimulation after 0.6 seconds.

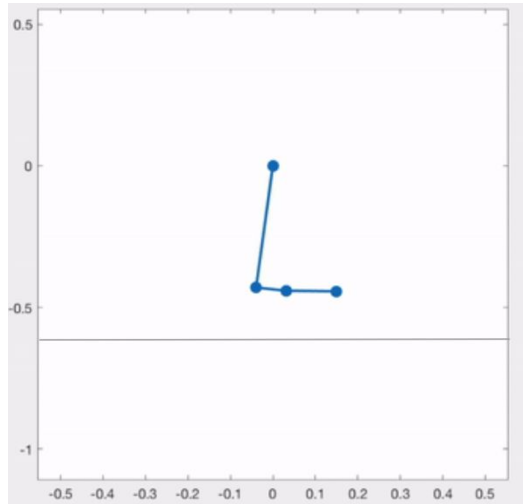


Figure 9b. Foot drop patient simulation with TA stimulation after 0.6 seconds.

Throughout these simulations, the contractile element (CE) length and resultant force were plotted against time as a means to visualize the TA activity during the swing phase. To do this, MATLAB's ode45 function was utilized with implicit variables for time, t, and normalized length of the contractile element, y. This function was then passed the stimulants produced through the activation model, aTA, in combination with the length parameter, L, as means to calculate the overall velocity of the CE. Note that the value of L was calculated to be 0.3314. The plots shown in **Figure 10a** and **Figure 10b** below are based on the ode45 solutions.

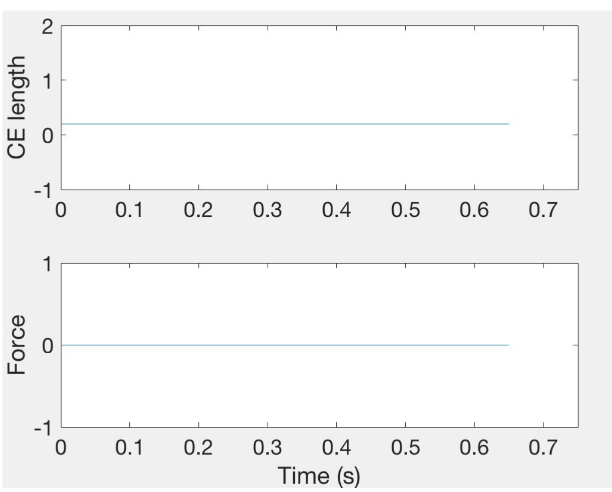


Figure 10a. CE length and force without TA stimulation.

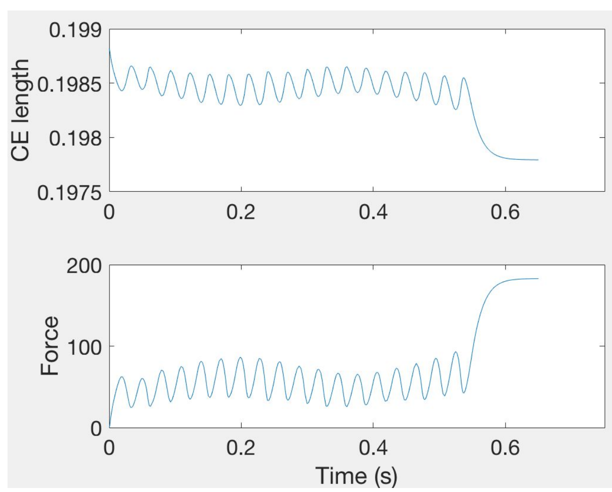


Figure 10b. CE length and force with TA stimulation.

Then, as verification for the activation stimulus applied, the toe height in metres was calculated as a function of both the shank and foot angles before being plotted against time.

The toe height with and without TA stimulation can be seen below in **Figures 11a** and **11b** respectively.

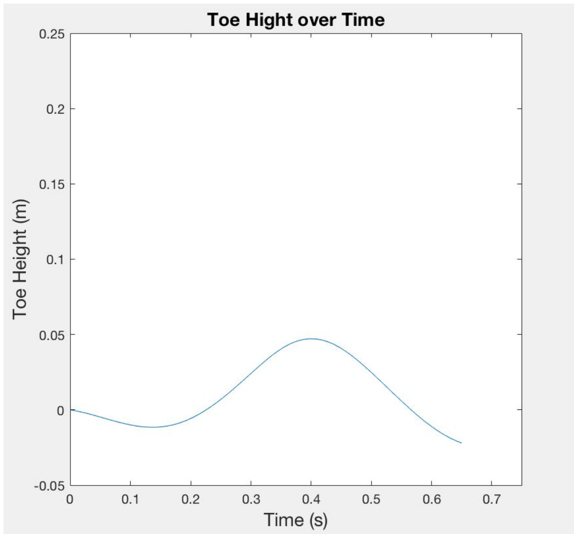


Figure 11a. Toe height without TA stimulation.

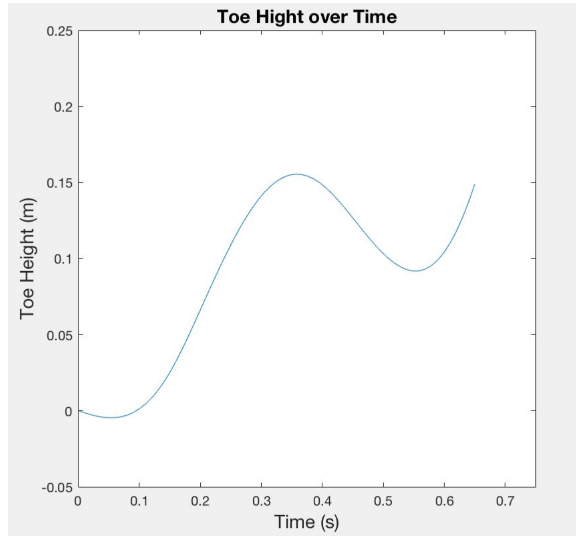


Figure 11b. Toe height with TA stimulation.

Furthermore, the result of incorporating a TA stimulation can be seen in the limb trajectories shown in **Figures 12a** and **12b** below. Here, the trajectories are shown about the fixed knee position at the origin (0,0) where the blue plot refers to the ankle and red, the foot's center of mass (COM).

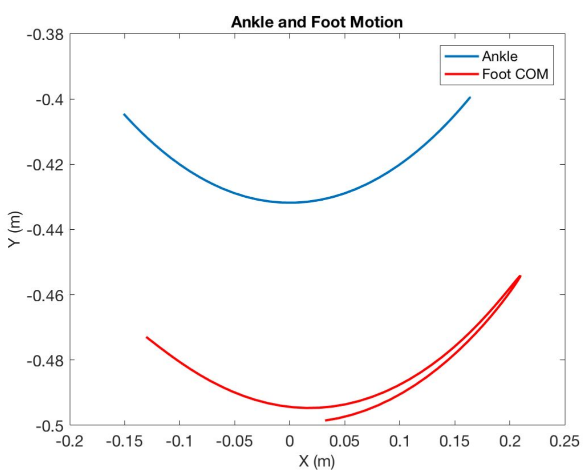


Figure 12a. Limb trajectory without TA stimulation.

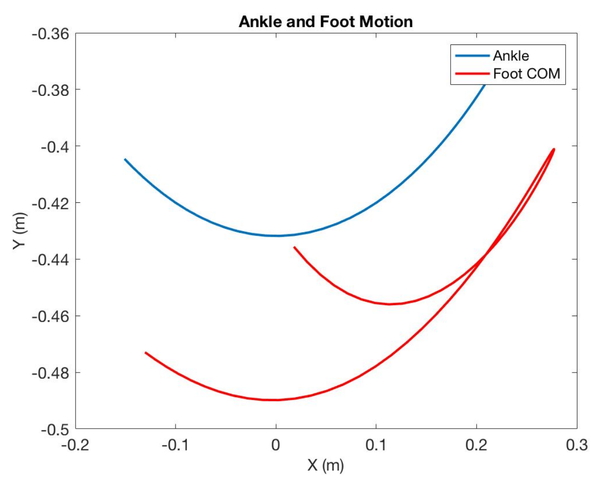


Figure 12b. Limb trajectory with TA stimulation.

5 Discussion

5.1 Interpretation of Results

Throughout model development, the key objectives were to understand and accurately capture the biomechanical dynamics of the foot and shank during the swing phase of the gait cycle. This in turn would fuel the end goal of establishing the electrical signal necessary to stimulate contraction of the TA muscle while simultaneously minimizing muscle fatigue, and determine whether or not that stimulation is sufficient to overcome drop foot.

When the muscle activation of the TA is zero, as shown in the bottom graph of **Figure 10a**, the elevation of the toe is found to remain close to the graphically-defined '0' or 'ground' reference, as seen in **Figure 11a** for the duration of swing phase. The toes were only above ground for half the duration of the swing phase of simulation one, from approximately 0.25 to 0.55 seconds. This reflects the initial position of the foot above the ground and its forward swinging motion as caused by the initial angular velocities applied to both the shank and the foot. However, during the approach to heel strike, during which the shank begins to swing backward, the toes are seen decreasing back to and below zero metres. The trajectory of the COM of the foot, as shown in **Figure 12a**, consequently remained near the 'ground'. Note that 'ground' in this instance refers to the lowest negative height shown in the figure. This behaviour is due primarily to the lack of change in the TA force which, according to the model definition, depends on the activation of the muscle. This force is then used to define the resultant CE length of the muscle which was found to remain constant at its normalized resting length of 0.1988, shown in the upper graph of **Figure 10a**.

These results are reasonable considering the model definition and biomechanics. A pendulum with no applied force to generate angular velocity about its fixed point will experience rotational translational oscillations that relax towards a stand-still position; i.e. the pendulum will swing until its mass hangs directly downwards. Translating this motion to the modelled swing phase, an individual would normally counteract this relaxation given the activation and contraction of the quadriceps; in doing so, they maintain forward motion of the first pendulum in this double-pendulular system. Additionally, during this forward motion the foot should be dorsiflexed due to activation and contraction of the TA muscle, controlling the second pendulum's range of motion. However, this pendular control is based on the TA's ability to generate enough tension to overcome both the gravitational forces as well as the foot's moment of inertia about the ankle joint. Without this, the foot is drawn downwards towards a relaxed position. As such, when the activation of the TA is zero in the first simulation, the muscle generates no force and does not induce dorsiflexion. Consequently, the toes are not raised high enough during swing phase and are dragged along the ground, as observed in **Figure 9a**. As a result, they are also incorrectly configured for the subsequent heel-strike. With this in mind, the model can be considered to accurately simulate the biomechanics of drop foot and, as such, can be used to determine the effectiveness of the applied electrical stimulus to the TA.

When the muscle activation of the TA was applied as a 34Hz oscillating stimulus, as shown in the bottom graph of **Figure 10b**, the toe elevation for the duration of swing phase was observed to rise and remain elevated above 'ground'. Note that the toes experienced a slight decrease in height in the first 0.1 seconds, corresponding to the initial forces acting on the foot prior to TA activation, though they quickly recovered. The toes maintained an average height of about 12.5 cm, as shown in **Figure 11b**, for the final 0.3 seconds of the swing phase. This path is similar to the shape of the EMG curve observed previously in **Figure 5** for the TA activation stimulus. Furthermore, the trajectory of the COM of the foot in **Figure 12b** concurs with these results, remaining consistently higher than 'ground' level when compared to that of the first simulation. This behaviour is the direct result of the oscillating TA CE length and force observed in **Figure 10b**. However, with an average generated force of 50N, the model is not consistent with literature to which forces measure in the hundreds of Newtons [24]. Despite this, the force does demonstrate the effect of the TA activation signal and one may conclude that this signal is sufficient to overcome the experience of drop foot.

Similar to the first simulation, these results are physiologically reasonable. Given the same initial pendular motion of both the shank and foot described previously, the application of an oscillating activation signal to the TA enables the muscle to overcome the gravitational forces and the foot's moment of inertia about the ankle joint. The resultant torque sustains the foot's angular velocity and rotation about the ankle, keeping the toes elevated above 'ground' as shown in **Figure 9b**. It is worth noting that the activation signal caused the CE length to significantly shorten near the end of swing phase, causing resultant TA force to jump to about 200N. This may be explained, however, by the necessary sharp dorsiflexion of the foot near the end of swing phase in preparation for the heel-strike. The toes, or distal end of the foot must not simply be elevated above ground, but elevated with respect to the heel. Additionally, having generated a large torque about the ankle, the increased TA tensile force primes the foot to counteract the opposing torque that will be generated as the heel strikes the ground. Considering this, the activation signal generated by the model sufficiently allows for an individual to overcome drop foot.

5.2 Conclusions

It was found that the application of an electrical stimulus to the TA at 34Hz, the optimal simulation frequency for minimizing muscle fatigue, was sufficient in shortening the TA CE length and thereby generate tensile forces comparable to those experienced normally. Using this stimulation, the elevation of the toes from the ground remained consistently higher when the TA had experienced electrical stimulation, compared to when there was no stimulation (indicative of drop foot). Similarly, the elevation of the COM of the foot was kept on an overall higher trajectory from the ground. In this way, as defined by the model, these results together allow for the conclusion that the application of a 34Hz electrical stimulation to the TA during swing phase is sufficient for an individual to mitigate the symptoms related to drop foot.

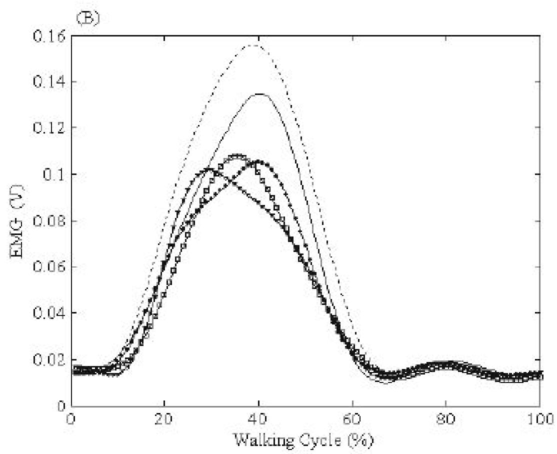
5.3 Limitations

For feasible development of the biomechanical swing phase model considering the resources and time available, assumptions that could affect the authenticity of the model were made. First of all, the activation of the soleus muscle was considered to be negligible during the swing phase, and hence the activation of the soleus muscle was assumed to be zero during this period. This, however, is not anatomically accurate and could potentially could affect the angular velocities and angles of the system, as the soleus generates an opposing moment to that of the TA. A second assumption was the presumption of the knee to be a fixed rotational joint around which the shank and the foot revolve - another anatomically incorrect assumption. This also affects the angular velocities and angles of the shank and ankle, as the motion of the knee and remainder of the body should, in reality, affect the motion of the lower leg. To further define the model, it was assumed the swing phase encompassed between 60 to 100% of the gait cycle; in reality, this percentage may vary between gait cycles. For the sake of simplicity, it was also assumed that the shape of the foot may be well-approximated by a scalene triangle; this inherently affects the calculated values used in the model, such as the height of the COM from the base of the foot, since various lengths are defined as percentages of anthropometric measurements. Finally, with regards to the Hill-type muscle model, the pennation angle and damping factor were assumed to be 0 and 0.1, respectively, for simplicity. Alterations to this value will affect the force and moment generated by the muscles in question.

5.4 Recommendations

Further analysis and model improvements can be made to ensure that the current model is an accurate representation of gait, and enable further understanding of how to optimally mitigate drop foot. To begin, the optimal stimulus frequency of 34Hz was chosen based on the stimulus with the least relative fatigue as seen in **Figure 4**. However, a stimulus with a greater frequency such as 46Hz could be used to produce greater activation given their proportional relationship. This increase could result in a better overall activation stimulus, possibly being more comparable to the normal activation of the TA while walking and more comfortable for the patient. This increased frequency, however, results in a slightly increased amount of muscle fatigue. Therefore, further testing is necessary to determine if the tradeoff between activation and fatigue can be further optimized in order to mimic normal TA activation more accurately. Future fatigue analysis should also incorporate the force produced by the TA throughout the swing cycle.

In addition, the current model only used the activation produced by the TA and assumes that the activation exhibited by the soleus is zero. Although the activation exhibited by the soleus during the swing cycle is minimal, taking the actual activation into account would result in a more anatomically accurate model. The gait cycle data in the Barela et al. research paper is found below in **Figures 13** and **14** [4]. Once again, the data of interest is defined by the solid black line.



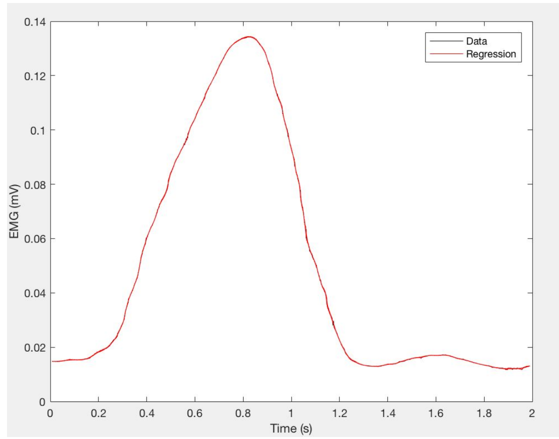


Figure 13. Mean trajectories of EMG from soleus during the walking cycle [4].

Figure 14. Regression modelled trajectories of EMG from soleus during the walking cycle.

To further increase the accuracy of the FES activation model, future iterations should include duty cycle as an optimizable parameter. The shape of the stimuli would remain as triangular peaks, however, the activation signal would not consist of constant peaks, but rather contain various stimuli separated evenly by a time during which no stimulus occurs. This would form a triangular pulse train, much like that depicted in **Figure 6**. The reason for this change would be to further decrease the fatigue of the TA muscle, avoiding a consistent activation of the muscle throughout the gait cycle. However, the time duration in which the activation is zero between peaks must be optimized, ensuring that the stimuli are not too close together causing fatigue, nor are they too far apart causing discomfort due to the unnatural time between peaks.

The implemented model should be compared to other existing models to identify any deviations. For example, the graphs and values generated by the developed model can be compared to those of other models to examine trends, particularly for the moment and angle of the body segments. Furthermore, to increase the comprehensiveness and accuracy of the model, the effect of other components of the body should be incorporated. For example, the motion and trajectory of the hip during gait will have an effect on those of the shank and the foot; additionally, other muscles in the lower leg such as the gastrocnemius and the extensor hallucis longus will also have an effect on the dynamics of the model.

Currently, the knee joint is assumed to be a fixed rotational joint, an anatomically inaccurate assumption; as such, another recommendation is to allow for this joint to have its own trajectory. In addition, to accurately model drop foot, frictional forces generated from the foot dragging on the ground during the swing phase should also be incorporated into the dynamics equation, as this will likely increase the activation required of the TA for the toes to clear the ground. Another method to increase the performance of the developed model is to provide better estimates for model parameter; for example, increasing the accuracy of measured anthropometric values.

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8 Appendix

A summary of the anthropometric values used throughout the biomechanical model is given below in **Table 1**. All angles are defined in radians, all masses in kilograms, and all distances in metres, and all lengths in metres.

Table 1. Summary of anthropometric values used in the definition of the biomechanical model.

Anthropometric Parameter	Value
Length of foot	0.2921 m
Length of shank	0.4318 m
Distance between knee and COM of shank	0.2133 m
Distance between ankle and COM of foot	0.07136 m
Distance between COM of foot and toes	0.117883 m
Angle between COM of foot and toes	1.43577 rad
Mass of shank	3.49431 kg
Mass of foot	1.10559 Kg