Evaluating a Muscle Reflex Model for Drop Foot Gait Simulation

Michael Dermksian[1], Alanna Mitchell[1], Vybhav Murthy[1], Eric Rasmussen[2]
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Abstract

Drop foot is a medical condition caused by the weakening of the Tibialis Anterior muscle. Here, we show that we can simulate gait patterns of patients with drop foot with a 2D muscular reflex model in order to potentially assist in diagnosing and treating patients. Seven Hill-type muscles are activated in the model to initiate and maintain gait. The overall isometric force of the Tibialis Anterior was decreased to 33% of its normal value in one leg to simulate drop foot. The joint angles from the drop foot simulation over the course of one stride were recorded and compared to experimental data, with discrete frechet distances of 10.923, 23.775, 8.427 for the ankle, knee, and hip respectively. The stride length, duration, and average velocity were 1.010m ± 0.126, 1.638s ± 0.052, and 0.650 m/s respectively; all values fall within one standard deviation of the experimental values.

Introduction

Legged locomotion of humans is complex and contains many layers of feedback pathways for control. Drop foot is a widely known phenomenon that usually occurs due to neurological damage or muscle weakening of the tibialis anterior [1]. In afflicted patients, compensatory mechanisms are activated in order to counteract the condition [2]. Compensatory mechanisms include recruiting other muscle groups to restore balance to human gait patterns. Typically, patients with foot drop tend to walk with an exaggerated flexion of the hip and knees to prevent their toes from dragging on the ground during the swing phase [3].

Biomechanists have exploited the use of muscle reflexes, the crux of legged dynamics, to develop a model for human locomotion [4]. Their application inspired us to utilize Geyer and Herr's 2D walking model in order to develop and optimize a model for drop foot: the motivation being, to be able to provide insight into a patient's injury state which could assist in diagnosis and treatment. The model was optimized with some of the basic ideas presented by Wang et. al's approach to optimizing the locomotion controllers with biological based actuators [5].

Model

Neural-MusculoSkeletal Model

In this project, the muscular reflex model produced by Geyer and Herr was used as the starting point [4]. The muscular reflex model utilizes a three layered approach consisting of a neural, muscular, and skeletal layer to produce locomotion when provided muscle stimulation.

The seven major muscles used during locomotion act as Hill-type muscles, and are given specific gain values and pre stimulations in the neural layer of the model. These stimulations are then inputted into the muscular layer, where they are used to generate muscle activation of each other their respective muscles. The muscle-tendon complex produces a max isometric force based on this activation, among other variables, and a muscle torque is generated based on the moment arm of the muscle and this isometric force. The muscle torques from each muscle are sent to the skeletal layer, where the signals act as an input to skeletal actuators,

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[1] Carnegie Mellon University, Department of Mechanical Engineering
[2] Carnegie Mellon University, Department of Biomedical Engineering

thus moving the body and producing locomotion. Positional data such as the timing of heel-strike/toe-off, and joint angles are sent back to the neural and muscular layers respectively, completing the feedback loop.

Drop foot patients lose muscle activity of the Tibialis Anterior (TA) in the affected leg at varying rates. Liu et al. performed a quantitative study on the muscle strength of the TA in patients with drop foot. His results showed that the average individual retained 33% of their TA strength compared to an uninhibited condition [5]. To simulate this disturbance, the max isometric force that can be produced by the TA in one leg was incrementally decreased and key parameters were optimized for at each interval. By the end of the boot-strapping process, our model was able to walk with a TA max isometric force of 265 N in the affected leg, 33% of the full strength condition of 800 N, which was maintained in the unaffected leg. By using this approach, we hope to compare the differences in our model and experimental data to examine the viability of this approach as a potential diagnostic tool.

Optimization

To create a model that resembles human drop foot data, our team utilized Covariance Matrix Adaptation-Evolution Strategy (CMA-ES) to optimize our affected model. The parameters used to conduct this optimization consisted of 16 variables for each leg (32 total). These parameters included muscle gains, and other preliminary gains associated with each leg such as offset and positional gains. The step size (σ) used to run the optimization was 0.0005 and the population size (λ) per generation was 12. The optimization aims to minimize a cost function which was written to have three terms: one for metabolic effort, one for target speed, and one for distance traveled [6], [7]. These terms are calculated as follows:

$$c_{effort} = \int_{o}^{T} \sum_{m \in M} a_m^2 \qquad c_{speed} = \int_{0}^{T} \left| 1 - \frac{v_x}{v_{tgt}} \right| \qquad c_{dist} = \frac{1}{d}$$

$$cost = w_{effort} c_{effort} + w_{speed} c_{speed} + w_{dist} c_{dist}$$

Weffort = 1, Wspeed = 2, Wdist = 100

Results

Gait patterns of the leg affected by drop foot were observed by recording the hip, knee, and ankle joint angles and torques while walking. Curves describing the angles of the ankle, hip, and knee were obtained by simulating the optimized model for 20s and separating out distinct gait cycles for the affected leg after the gait has stabilized. Time was then normalized to a percentage of the gait cycle, and all curves were averaged. In figures 1-3, this average curve is represented in blue. The figures below show the gait pattern graphs generated for optimized runs at a tibialis anterior maximum isometric force of 265 N.

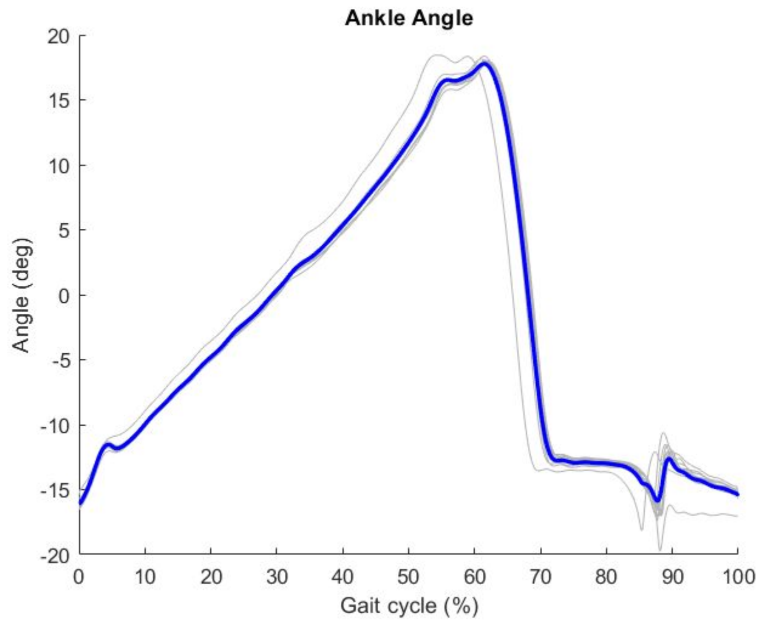

Figure 1: Average Ankle Angle
The above figure shows the trajectory of the ankle joint over a single stride. The blue data line represents the average ankle angle, while the faded gray lines represent a single stride.

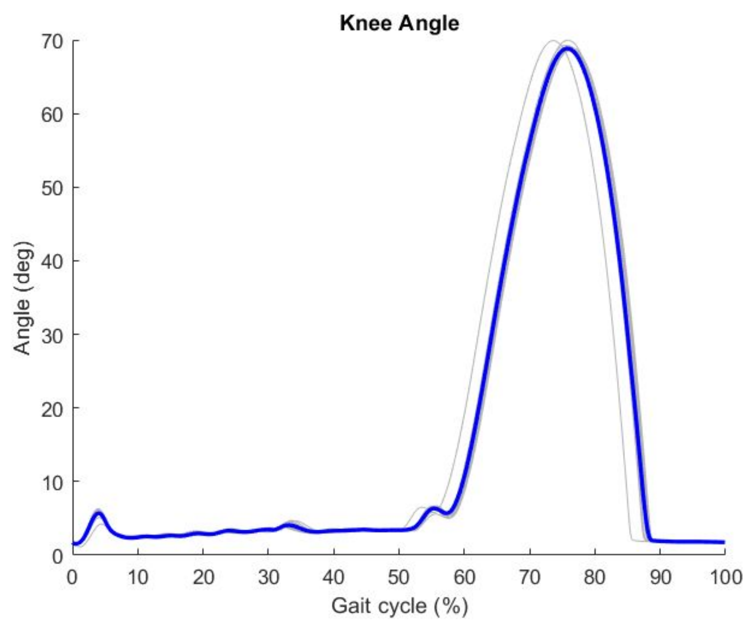

Figure 2: Average Knee Angle
The above figure shows the trajectory of the knee joint over a single stride. The blue data line represents the average knee angle, while the faded gray lines represent a single stride.

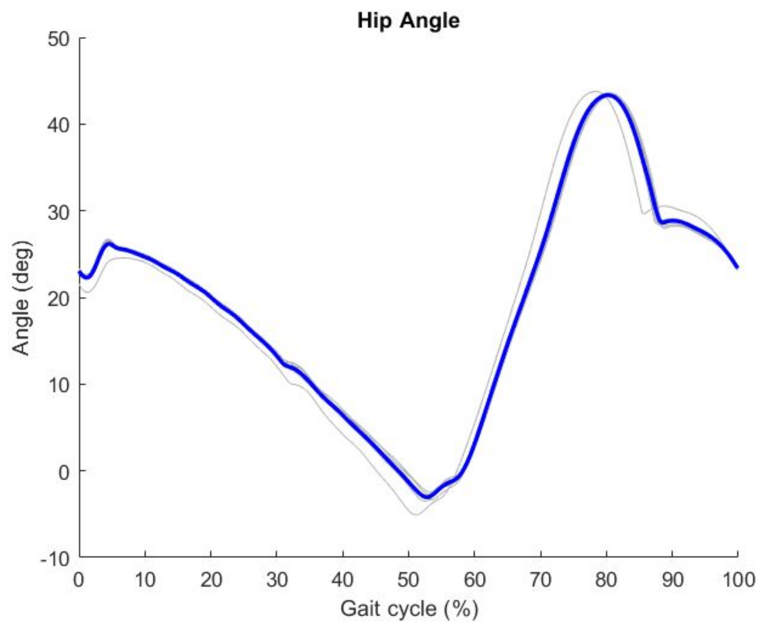

Figure 3: Average Hip Angle

The above figure shows the trajectory of the hip joint over a single stride. The blue data line represents the average hip angle, while the faded gray lines represent a single stride.

The simulated model had a stride duration of 1.638s ± 0.052, a stride length of 1.010m ± 0.126, and an average velocity of 0.650 m/s.

Discussion

The results of this project were compared to an experimental study conducted by Wiszomirska et al. on patients with drop foot caused by degenerative disc disorder [2]. A study of surgical outcomes of drop foot patients with lumbar degenerative diseases used a manual muscle test to quantify the strength of the tibialis anterior. The average muscle strength of these patients was 1.66 on a scale of 0 to 5, or 33.2% [5]. Based on this data, the maximum isometric force of the tibialis anterior was reduced from 800 N to 265 N on the affected drop foot leg. Normalized and averaged gait pattern graphs for the hip, knee, and ankle angles and torques were compared to those obtained in Wiszomirska's study [2].

The joint angle data from the results section was compared to experimental curves presented by Wiszomirska, et al for the equivalent angles, represented in red and black. The simulated joint angles are still represented in blue. These graphs can be seen in the appendix. To perform this comparison, discrete Fréchet distances were taken between the simulated curve and the experimental drop foot curve. Fréchet distances were also taken between the simulated curve and the experimental control (healthy) curve.

Frechet Distances	Simulated - Experimental DF	Simulated - Experimental C
Ankle Angle	10.923	17.057
Knee Angle	23.775	16.033
Hip Angle	8.427	35.546

Optimized simulation data for hip and ankle angles are closer to the experimental data Wiszomirska found for drop foot patients. However, the knee angle data are close to the healthy patients. This may be because real drop foot patients are capable of moving in three dimensions, and may reduce the need to flex the knee by swinging the leg outwards in the sagittal plane. This compensation is impossible for our 2D model, so it must flex the knee to prevent toe stubbing.

Several spatiotemporal parameters were compared between the simulated model and drop foot subjects from Wiszomirska's study. Our simulated model had a stride duration of 1.638s ± 0.052, a stride length of 1.010m ± 0.126, and an average velocity of 0.650 m/s. Wiszomirska reports a stride duration of 1.7s ± 0.4, a stride length of 1.1m ± 0.1, and a walking speed of 0.7m/s ± 0.1. All simulation parameters fall within one standard deviation of the experimental means.

Future work on this project should come in the form of adding a third dimension to the simulation. As previously mentioned, our model is not allowed to compensate for its condition in the sagittal plane due to its 2D restriction. This leads to overcompensation by raising the knee angle higher than it should suggested by experimental data. Investigation into this matter could lead to a closer comparison between or model and experimental data.

Citations

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Appendix

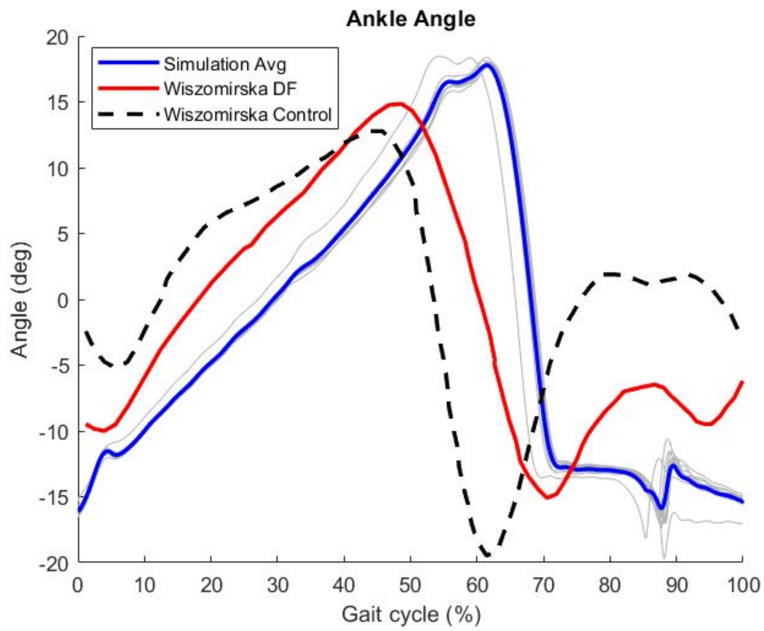

Figure 4: Ankle Angle Comparison
The above figure is the same as figure 1, with the experimental data overlaid for comparison.

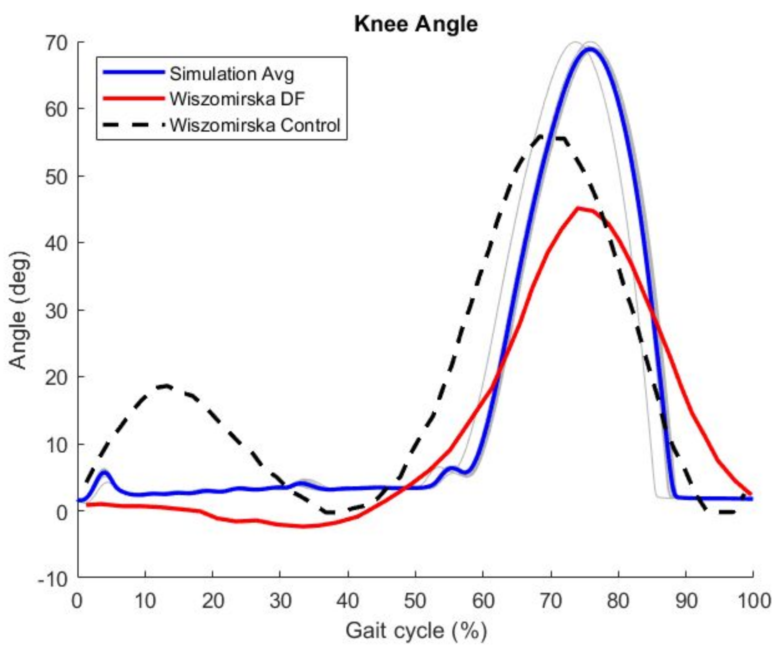

Figure 5: Knkle Angle Comparison
The above figure is the same as figure 2, with the experimental data overlaid for comparison.

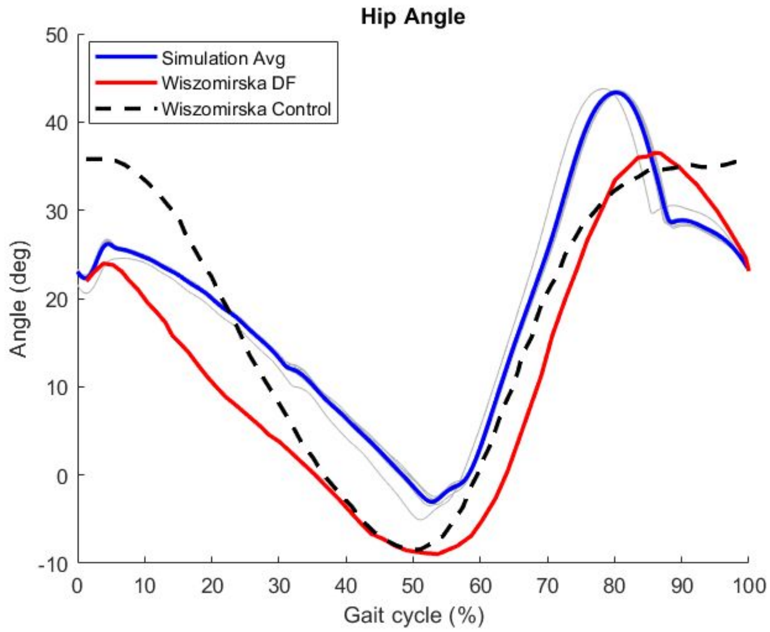

Figure 6: Hip Angle Comparison
The above figure is the same as figure 3, with the experimental data overlaid for comparison.