Virtual Design of an Assistive Stair Ascent Device for Individuals with Knee Osteoarthritis

Undergraduate Honors Thesis

Presented in Partial Fulfillment of the Requirements
for Graduation with Honors Distinction in the
Department of Mechanical and Aerospace Engineering at
The Ohio State University

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December 2020

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Abstract

Osteoarthritis (OA) is a disease that is becoming increasingly more prevalent due to heavier and aging populations. Effects of OA in the knee range from weakness and instability to loss of range of motion, resulting in difficulty performing everyday tasks such as walking, stair negotiation, and chair rise. With few options to mitigate these effects, individuals with knee OA may turn to noninvasive devices in an attempt to decrease pain and increase function while taking on these everyday activities. There are currently few devices available to such individuals to assist in stair negotiation, most of which are expensive or ineffective. Additionally, studies show that individuals with knee OA biomechanically climb stairs differently, due to the compromised joint. However, no devices on the market are currently designed for individuals with knee OA. The goal of this project is to simulate an assistive device that aids individuals with knee OA in stair ascent.

The Neuromuscular Biomechanics Research Laboratory previously collected motion data of individuals with knee OA during stair ascent. In this study, I used a program called OpenSim Moco to create a model and run dynamic simulations. I simulated an assistive device by placing torsional springs of varying stiffnesses at the hip, knee and ankle to observe how stair climbing was affected. The goal was to simulate lower limb kinematics of those of a healthy individual, without increasing metabolic cost. Results from this study showed that no spring placement or stiffness had a positive impact on both metabolic cost and maximum muscle force.

The results of the simulations from this study and future work can provide insight for design parameters for an assistive stair climbing devices for individuals with knee OA, and possibly improve their performance in daily activities.

Acknowledgements

First, I would like to thank my advisor, Dr. Rob Siston. I have had the pleasure of working with him for the past two and a half years and he has shaped so much of my undergraduate career. He has pushed me to explore new ideas and pursue new projects, which exposed me to topics and ideas I would have not otherwise learned about. He has provided so much encouragement and support, and I am so grateful for him.

I would also like to thank the other NMBL members, especially Anna Lee and Rachel Hall for the support they provided on working with OpenSim. Thank you also to the other lab members for their feedback and help.

Finally, thank you to my family and friends. Though they may not have understood what all my thesis was, they have been so encouraging and I am incredibly grateful for their support.

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Chapter 1: Introduction

1.1 Background

Individuals with Knee Osteoarthritis

Nearly 27 million adults in the US have clinical osteoarthritis (OA), with knee OA being one of the most common forms [1]. This number is expected to grow due to occurrence of OA being closely related to increasing age and obesity [2]. Knee OA is defined as the breakdown of cartilage that narrows the joint space and results in bone-on-bone contact [3]. Effects and severity of knee OA can vary across individuals; however, the most common symptoms include pain, weakness, swelling, and stiffness [4]. These symptoms can then cause further difficulty with mobility due to decrease or loss of range of motion. In turn, individuals with knee OA can experience difficulty in completing everyday tasks such as walking, kneeling, squatting, or stair negotiation [5].

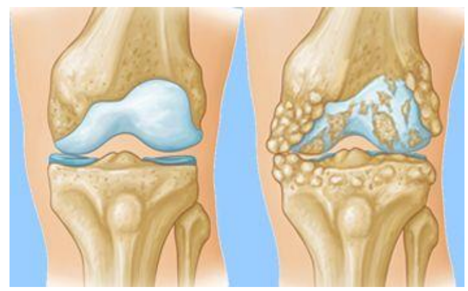

Figure 1: Healthy vs. Osteoarthritic Knee [3]

Stair Ascent and Compensation with Knee OA

Stair ascent is an essential daily activity that many individuals with knee OA struggle with. Stair ascent relies heavily on the quadriceps muscles, which are typically weakened and can act as a precursor to knee OA. Studies have found that individuals with knee OA exhibit muscle atrophy and activation deficits that result in overall quadriceps weakness [6]. Because of this, they adopt compensatory mechanisms to allow them to climb stairs. These mechanisms present themselves in a variety of biomechanical differences. Individuals with severe knee OA have an increased forward trunk lean to reduce demand on the quadriceps [7]. Additionally, it has been found that these individuals compensate with decreased peak knee flexion and extension moments, an increased peak hip flexion moment, and an increased varus moment [7, 8]. It is possible that these compensatory patterns could be detrimental to other joints because of the altered biomechanical loading. Achieving or restoring normal joint mechanics could be beneficial for both the affected knee and remaining joints [7].

Current Stair Ascent Devices on the Market

There are currently few devices on the market that aid individuals in stair ascent (Figure 2). The EZ-Step Cane (Figure 2A) is a freely mobile device that aids in stair ascent by decreasing the vertical distance to be traveled, therefore decreasing the required hip and knee range of motion [9]. However, the cane relies on the ability of the user to safely and effectively maneuver it. Another option is a stair lift (Figure 2B), which eliminates the need to physically climb stairs altogether. Though this is easy to use for the user, it must be installed on any set of stairs the user wants to negotiate and can cost thousands of dollars [10]. The Honda Walking Assist (Figure 2C) is a device worn around the user's waist and thighs that uses motors to power the legs forward. The device weighs around 6 pounds and one charge can only last 60 minutes [11]. Additionally, it is not available for purchase globally. When the product was first available to lease in Japan, it cost $362 per month [12].

Figure 2: Stair Ascent Assistive Devices

A device that is specifically marketed to individuals with knee OA is an unloader brace (Figure 2D). Unloader braces are designed for individuals with knee OA in one compartment of their knee. Unloader braces use a 3-point loading system to shift the force seen in the knee to the unaffected compartment and offload the affected compartment. Although this device is relatively light and mobile, it requires the user to have only unicompartmental knee OA and does not actually provide any mechanical assistance for stair ascent [13]. Though all of these device options can provide aid in stair ascent, none of them specifically target the altered biomechanics of individuals with knee OA.

Using Simulations to Predict Movement

Simulations of human movement can be used as a powerful tool to provide qualities or parameters that cannot be determined experimentally, like muscle activations or metabolic cost. OpenSim is an open source software that allows the user to model and study human movement across tasks like gait or stair climbing [14-16]. OpenSim Moco allows researchers to perform simulations to estimate how muscles in the lower extremities could respond to different forms of assistance from a device by solving predictive simulations using a solving method called direct collocation. Using simulations in OpenSim is a way to test many variations of virtual devices and view predicted outcomes with very little time and resources, compared to building and testing a physical device.

For example, one study was performed using OpenSim to simulate movement of an elderly adult while performing activities of daily living (ADL), including gait, stair ascent, and stair descent [17]. The goal was to determine actuation requirements in order to create a low-profile exoskeleton for both hip and knee assistance. OpenSim was used to simulate movement for gait, stair ascent, and stair descent and provided torque values for the hip and knee that fed into design parameters. A study similar to this is needed to determine design parameters for an assistive stair ascent device for individuals with knee OA. The goal of this device is to aid in stair ascent by decreasing necessary muscle force and metabolic cost, not to rehabilitate the muscles in hopes of restoring previous function without the device.

1.2 Focus of thesis

The focus of this project is to virtually design an assistive device for individuals with knee OA, based on their compensated biomechanics, and evaluate the corresponding metabolic cost and muscle forces.

1.3 Significance of Research

Though there are options on the market that aid in stair ascent, none of the devices are designed specifically for individuals with knee OA. This is a very large and high need population that needs to be addressed with an assistive device. This study will investigate design parameters for a wearable assistive device to provide guidance for physical design and prototyping to aid individuals with knee OA during stair ascent.

1.4 Overview of Thesis

This thesis contains four chapters. The second chapter will present the methods of creating the model and running simulations of the assistive device. The third and fourth chapters will present and analyze the results of the simulations regarding the assistive device, and provide suggestions for future work.

Chapter 2: Methods

2.1 Experimental Data

The representative subject data was selected from a study of 23 older healthy individuals who provided written informed consent in accordance with the Institutional Review Board of The Ohio State University to participate in the study. The data were collected by Dr. Elena Caruthers and Dr. Sarah Roelker in the Motion Analysis and Performance Lab (The Ohio State University). All subjects had no known lower limb or nervous system pathologies. Data from an older healthy adult was used due to data accessibility issues with COVID-19.

Stair Ascent Motion Data

Experimental stair ascent (SA) motion data was collected by Dr. Elena Caruthers and Dr. Sarah Roelker. Custom force plates were installed under the first two stairs of a custom staircase (tread depth: 25.5 cm, step height: 20 cm) (Figure 3).

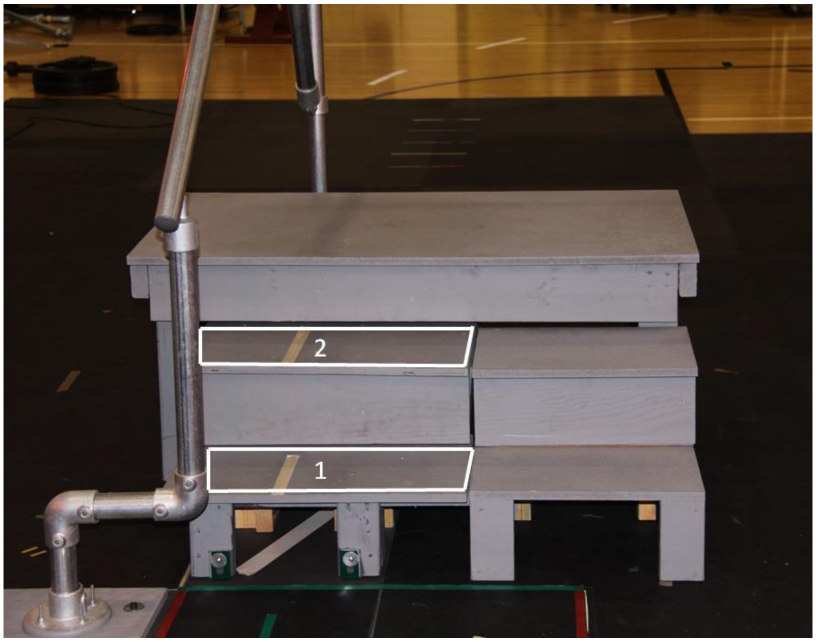

Figure 3: Instrumented stairs for data collection

Motion data were collected at 150 Hz using a 10-camera Vicon system. The subjects were tracked with a modified point cluster marker system (Figure 4).

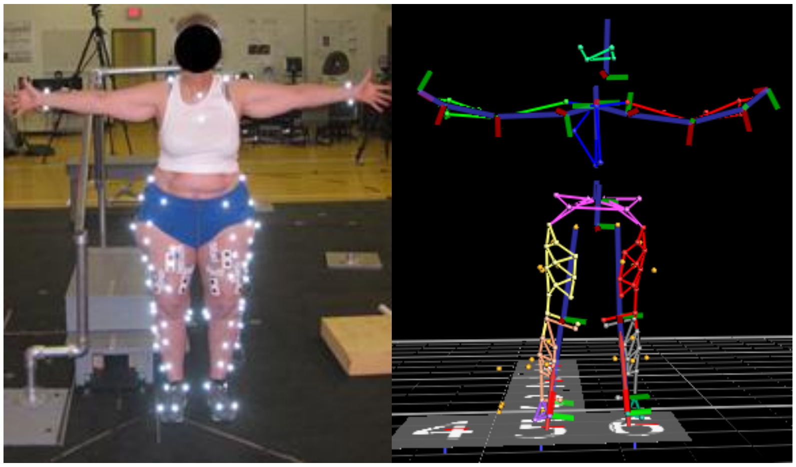

Figure 4: Modified full-body point cluster technique

Ground reaction forces were obtained through three force plates sampled at 600 Hz. The subjects completed 6 consecutive stair navigation trials at a self-selected speed. A representative SA trial was selected for this study (female, height = 1.74 m, weight = 59.60 kg, age = 57 years).

2.2 Musculoskeletal Modeling

I modified an example model that was a 10 degrees of freedom and 80 muscles OpenSim Moco armless model designed for walking with Millard 2012 Equilibrium Muscles. Three degrees of freedom – lumbar extension, rotation, and bending – were added to allow for a movable back. The model was then simplified to only twelve muscles per leg by combining groups of muscles to a single, representative muscle to reduce the number of variables to be solved for in the simulation. The muscles used were the biceps femoris, a reduced gastrocnemius, a reduced gluteus maximus, a reduced gluteus medius, a reduced gluteus minimus, psoas major, rectus femoris, a reduced semitendinosus, soleus, tibialis anterior, and a reduced vastus intermedius. To incorporate a characteristic of knee OA stair ascent, the quadriceps peak isometric muscle forces were reduced by 26.4% from their original values [18].

2.3 OpenSim Moco: MocoTrack

MocoTrack allows the user to input a model, motion data, and ground reaction forces, while also allowing the simulation to deviate from prescribed kinematics. Typically, tracking problems can be time consuming and difficult. However, Moco uses a solving method called direct collocation to simplify this computation. Direct collocation takes the initial and final states of a point, in this case a marker, and fills the movement in between with a spline approximation. The program then evaluates the dynamics function of the system at all intermediate points simultaneously to solve for error between the derivative of the spline approximation and the result of the dynamics function. Moco iterates through this process at all points in time until the error is within acceptable range and the simulated motion is achieved.

2.4 Torque and Muscle Driven Marker Tracking

Torque driven marker tracking (TDMT) and muscle driven marker tracking (MDMT) were used to provide baseline results for the simulations. A musculoskeletal model, ground reaction forces, and marker trajectory data were loaded into TDMT. The program then took the model, removed all muscles, and added reserve actuators of 200 Nm at each joint to track the motion and solve for required torques. MDMT takes the same inputs, but solves for necessary muscle activity to achieve the resulting motion based on marker data. These functions were run on the original model with no modifications to obtain the baseline data. These functions were then repeated on the OA model with weakened quadriceps for an additional baseline. The joint angle states from these results were taken and used as the input boundary conditions for the predictive simulation.

2.5 Assistive Device Simulations

Modeling Joint Assistance

For the predictive simulations, ideal, massless, torsional springs were applied to the ankle, knee, and hip independently. The stiffness was varied from 1 to 5 Nm/deg in 1 Nm/deg increments. Additional simulations were run with a 10 Nm/deg torsional spring placed on the ankle, knee, and hip independently. This resulted in a total of 18 simulations. The hip spring assisted in hip extension at 50 degrees of flexion. The knee spring assisted in knee extension at 70 degrees of flexion. The ankle spring assisted in plantar flexion at 5 degrees of dorsiflexion.

Predictive Simulations

The modified models with the springs were then run through an additional simulation to predict resulting motion and necessary muscle activation. The first predictive simulation attempted took the results of the muscle driven marker simulation as a guess for the resulting motion, and tried solving for the muscle activations needed to achieve the motion. This code was ultimately unsuccessful, presumably due to too many degrees of freedom, so a new predictive simulation that used joint angle boundary conditions and a similar, simplified model was used. The model had only one limb and 9 muscles to limit degrees of freedom, and had a weld joint attaching the foot to the floor. The muscles included were the biceps femoris, medial gastrocnemius, gluteus maximus, psoas, rectus femoris, semimembranosus, soleus, tibialis anterior, and vastus intermedius. Degrees of freedom were limited to hip, knee, and ankle flexion and extensions, with lumbar extension, bending, and rotation. This predictive code took joint angle conditions from MDMT as input and simulated one step being taken with the right leg from flat foot contact to full extension. Baseline tests were repeated for normal and weakened models. A MocoControlGoal was included that calculated the metabolic cost associated with the completed motion (Equation 1) – where x_c, the control signal of each muscle, was squared and summed at each time point over the trial.

$$\int_{t_i}^{t_f} \sum |x_c(t)|^2 dt \qquad \text{(Equation 1)}$$

3. Results

The results will be presented as follows: overall metabolic cost, individual muscle metabolic cost, maximum muscle forces, and joint kinematics. All values will be presented as percent changes relative to the baseline and weakened models.

3.1 Metabolic Cost for Predictive Simulations

Overall Metabolic Cost

The overall metabolic cost compared to the weakened model increased for all knee spring, one ankle spring, and two hip springs (Table 1). The ankle spring with stiffness k = 10 Nm/deg and hip springs of k = 3 and 10 Nm/deg increased overall metabolic cost, while the remaining ankle and hip springs decreased overall metabolic cost. The metabolic cost change for the ankle and hip springs did not occur in a uniform way with increasing stiffness, which could be an error in the simulation technology. The hip spring with a stiffness of 2 Nm/deg decreased the metabolic cost the most by 13.32% The most metabolically expensive device was the knee spring with a stiffness of 10 Nm/deg.

Table 1: Percent change in overall metabolic cost compared to weakened model

Joint	Stiffness (Nm/deg)	Percent Change
Ankle	1	-0.15
	2	-1.49
	3	-1.44
	4	-1.37
	5	-1.37
	10	0.10
Knee	1	4.10
	2	5.54
	3	11.24
	4	19.15
	5	29.31
	10	194.22
Hip	1	-11.79
	2	-13.32
	3	0.50
	4	-5.99
	5	-2.28
	10	2.01

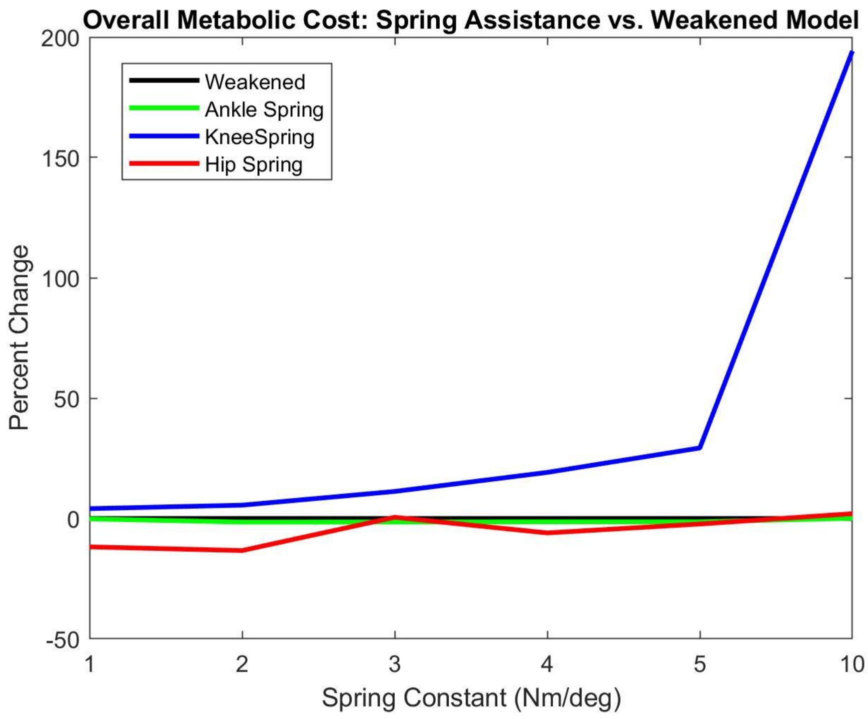

Figure 5: Percent change in overall metabolic cost compared to weakened model

Metabolic Cost for Individual Muscles

For individual muscle analyses, the first 17% of the data was excluded to remove any outliers. The first 17% of the step would have double leg support, so that data in this series is not representative of a real stair ascent cycle [19]. Metabolic cost for the individual muscles compared to the weakened model varied in a different way than overall metabolic cost. When the ankle was assisted, metabolic cost decreased for the rectus femoris and vastus intermedius, and metabolic cost increased for the biceps femoris, medial gastrocnemius, psoas, soleus, and tibialis anterior (Table 2). Change in metabolic cost for the gluteus maximus and semimembranosus varied depending on the stiffness of the spring. The muscle that benefitted the most from assistance at the ankle was the vastus intermedius.

Table 2: Percent change in muscle metabolic cost of ankle assistance compared to weakened model

Stiffness (Nm/deg)	1	2	3	4	5	10
Biceps Femoris (Short Head)	22.40	11.93	8.59	17.03	17.06	1.68
Medial Gastrocnemius	689.11	173.24	177.37	222.24	225.47	13.77
Gluteus Maximus	198.88	68.38	16.18	70.78	79.93	-27.21
Psoas	52.45	11.92	11.76	9.09	9.38	0.84
Rectus Femoris	-18.75	-4.31	-4.38	-7.93	-8.33	-0.03
Semimembranosus	-99.99	26.63	-23.61	70.43	66.89	2.94
Soleus	581.32	124.28	127.30	174.53	179.00	9.99
Tibialis Anterior	23.55	10.53	10.88	20.32	20.42	0.21
Vastus Intermedius	-45.56	-27.65	-27.70	-28.31	-28.53	-0.64

When the knee was assisted, metabolic cost increased for the medial gastrocnemius, rectus femoris, and vastus intermedius, while metabolic cost decreased for the biceps femoris (Table 3). The metabolic cost for the biceps femoris decreased by 100% for every spring stiffness. Change in metabolic cost varied for the gluteus maximus, psoas, semimembranosus, soleus, and tibialis anterior based on spring stiffness.

Table 3: Percent change in muscle metabolic cost of knee assistance compared to weakened model

Stiffness (Nm/deg)	1	2	3	4	5	10
Biceps Femoris (Short Head)	-100.00	-100.00	-100.00	-99.99	-100.00	-100.00
Medial Gastrocnemius	29.23	486.00	1070.90	2261.14	4025.65	25.32E4
Gluteus Maximus	-78.47	14.31	126.09	-36.72	2.13	9038.74
Psoas	-25.18	-26.58	-27.41	-32.42	-36.86	58.06
Rectus Femoris	21.85	26.18	44.43	60.83	78.89	1177.65
Semimembranosus	-26.93	-42.08	179.05	32.43	10.72	-99.99
Soleus	-82.82	-41.04	-2.02	133.49	374.25	35.85E3
Tibialis Anterior	-7.46	18.71	18.99	8.83	6.17	251.58
Vastus Intermedius	33.23	1.26	29.10	84.10	143.48	467.57

When the hip was assisted, metabolic cost increased for the medial gastrocnemius, gluteus maximus, and soleus, while metabolic cost decreased for the psoas and rectus femoris. The metabolic cost for the biceps femoris, semimembranosus, tibialis anterior, and vastus intermedius varied depending on spring stiffness.

Table 4: Percent change in muscle metabolic cost of hip assistance compared to weakened model

Stiffness (Nm/deg)	1	2	3	4	5	10
Biceps Femoris (Short Head)	18.11	-62.33	136.04	-87.99	-86.46	-99.85
Medial Gastrocnemius	373.87	750.77	2484.49	1873.36	3060.58	7704.52
Gluteus Maximus	3033.05	4926.80	5923.16	6557.74	12.24E3	44.23E3
Psoas	-92.25	-97.94	-94.71	-95.68	-69.67	-99.83
Rectus Femoris	-93.25	-87.65	-95.55	-84.37	-85.72	-99.99
Semimembranosus	-30.75	1820.93	2039.37	2502.88	4579.88	17.85E3
Soleus	314.23	589.66	1846.41	1109.18	1745.96	4317.76
Tibialis Anterior	10.81	-7.94	46.36	49.89	-23.31	-40.45
Vastus Intermedius	-31.58	-13.70	-65.77	-71.85	19.48	-8.60

Overall, all springs impacted metabolic cost of individual muscles in varying ways (Table 5). The medial gastrocnemius increased in metabolic cost with assistance at all joints.

Table 5: Summary of change in muscle metabolic cost for all joints and all spring stiffnesses

Joint	Increased Cost	Decreased Cost
Ankle (plantarflexion)	Biceps Femoris, Medial Gastrocnemius, Psoas, Soleus, Tibialis Anterior	Rectus Femoris, Vastus Intermedius
Knee (extension)	Medial Gastrocnemius, Rectus Femoris, Vastus Intermedius	Biceps Femoris
Hip (extension)	Medial Gastrocnemius, Gluteus Maximus, Soleus	Psoas, Rectus Femoris

3.2 Muscle Forces

Maximum Forces Compared to Baseline Model

When the ankle was assisted, maximum forces in the biceps femoris, medial gastrocnemius, and rectus femoris decreased while maximum forces in the tibialis anterior and vastus intermedius increased (Table 6). Maximum forces in the gluteus maximus, psoas, semimembranosus, and soleus varied based on spring stiffness. Maximum forces for the medial gastrocnemius decreased the most of all muscles. Muscle forces were plotted against baseline model for all spring stiffnesses in Figure 6. There is a ripple that is consistently seen in the data, which may be due to the simulation mesh size being only 30 intervals.

Table 6: Percent change in maximum muscle force of ankle assistance compared to baseline model

Stiffness (Nm/deg)	1	2	3	4	5	10
Biceps Femoris (Short Head)	-54.84	-55.77	-55.04	-54.83	-55.17	-56.00
Medial Gastrocnemius	-60.77	-73.68	-73.50	-72.46	-72.35	-72.88
Gluteus Maximus	-0.14	2.98	11.52	2.10	2.16	0.38
Psoas	12.00	-0.68	-1.02	-8.67	-9.09	-12.79
Rectus Femoris	-41.29	-29.27	-29.45	-33.41	-33.31	-32.41
Semimembranosus	20.04	5.00	4.91	31.55	31.07	-0.80
Soleus	30.27	-34.38	-33.69	-24.76	-24.28	-56.83
Tibialis Anterior	24.54	11.42	11.58	15.55	15.55	2.12
Vastus Intermedius	6.29	14.50	14.54	15.13	15.14	8.34

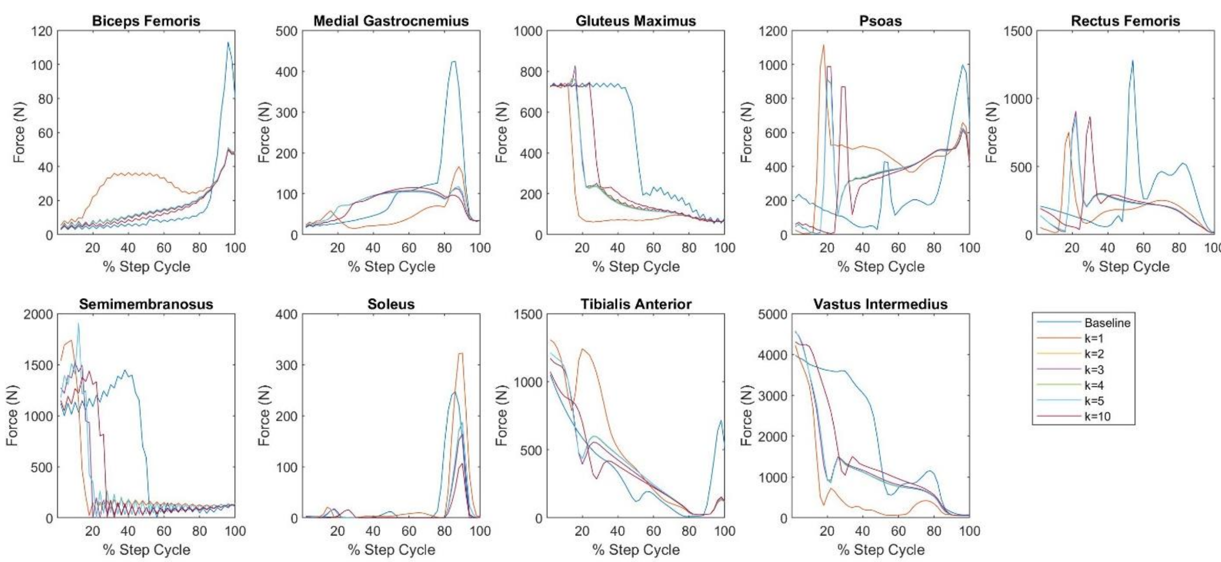

Figure 6: Muscle forces of ankle spring assistance compared to baseline model

When the knee was assisted, maximum muscle forces decreased for the biceps femoris and psoas (Table 7). Maximum muscle forces increased for the vastus intermedius. Change in maximum muscle forces varied for the medial gastrocnemius, gluteus maximus, rectus femoris, semimembranosus, soleus, and tibialis anterior. The knee spring with stiffness k = 10 Nm/deg had the largest impact on increasing maximum muscle forces, as exhibited for the medial gastrocnemius and soleus. Maximum muscle forces for the medial gastrocnemius, rectus femoris, and soleus all decreased for spring values 1 through 5 Nm/deg.

Table 7: Percent change in maximum muscle force of knee assistance compared to baseline model

Stiffness (Nm/deg)	1	2	3	4	5	10
Biceps Femoris (Short Head)	-69.13	-68.94	-68.88	-68.82	-68.75	-68.65
Medial Gastrocnemius	-71.94	-71.68	-67.53	-58.60	-49.25	188.05
Gluteus Maximus	-0.97	1.94	15.55	10.19	6.49	19.31
Psoas	-22.11	-17.34	-8.45	-7.17	-11.90	-47.75
Rectus Femoris	-33.75	-33.98	-27.18	-26.53	-27.47	14.06
Semimembranosus	-6.26	14.71	38.27	-2.85	-4.53	-65.96
Soleus	-89.25	-74.69	-56.92	-31.94	-2.59	708.88
Tibialis Anterior	-2.29	14.50	24.81	8.19	1.67	43.39
Vastus Intermedius	5.81	14.18	12.78	13.52	9.43	48.40

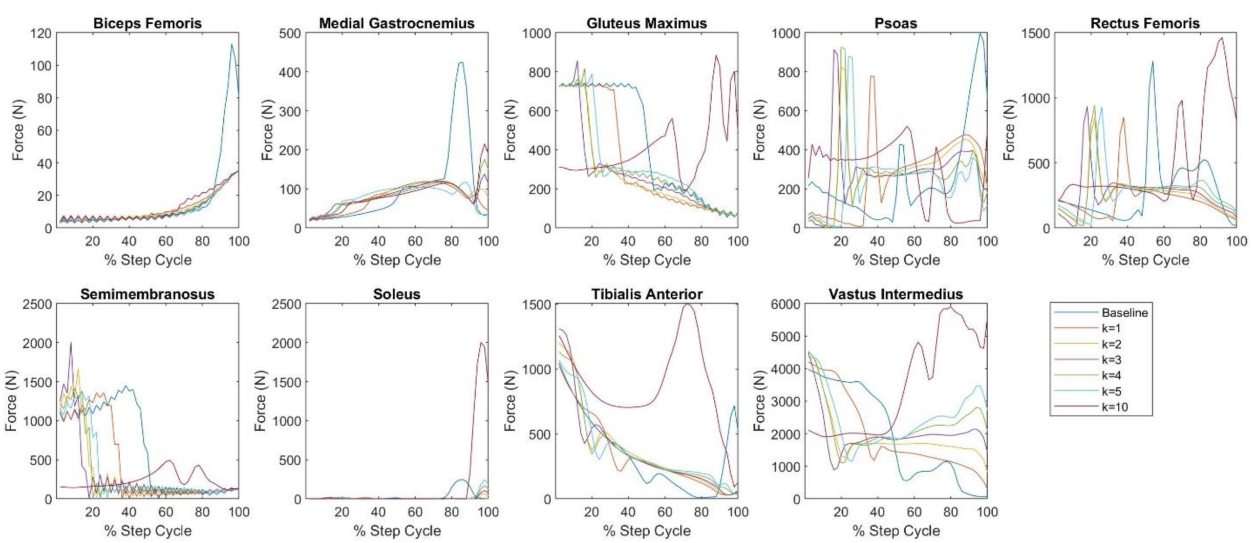

Figure 7: Muscle forces of knee spring assistance compared to baseline model

When the hip was assisted, maximum muscle forces increased for the soleus, while maximum muscle forces decreased for the biceps femoris, medial gastrocnemius, psoas, and rectus femoris (Table 8). Change in maximum muscle forces varied with spring stiffness for the gluteus maximus, semimembranosus, tibialis anterior, and vastus intermedius. The hip spring with the stiffness k = 10 Nm/deg caused a large increase for the gluteus maximus and semimembranosus.

Table 8: Percent change in maximum muscle force of hip assistance compared to baseline model

Stiffness (Nm/deg)	1	2	3	4	5	10
Biceps Femoris (Short Head)	-56.42	-63.86	-57.71	-65.02	-65.02	-68.27
Medial Gastrocnemius	-60.96	-57.38	-39.75	-58.29	-38.59	-27.44
Gluteus Maximus	-1.33	8.21	26.85	8.86	40.05	185.71
Psoas	-65.38	-90.28	-83.36	-85.99	-75.39	-96.03
Rectus Femoris	-90.80	-93.07	-80.59	-57.58	-79.09	-92.24
Semimembranosus	13.77	76.33	-9.29	-26.78	26.59	105.29
Soleus	23.77	93.16	155.57	77.41	133.76	70.06
Tibialis Anterior	16.93	1.11	36.24	50.64	6.53	-1.20
Vastus Intermedius	12.28	12.46	-37.01	-44.10	14.99	-22.42

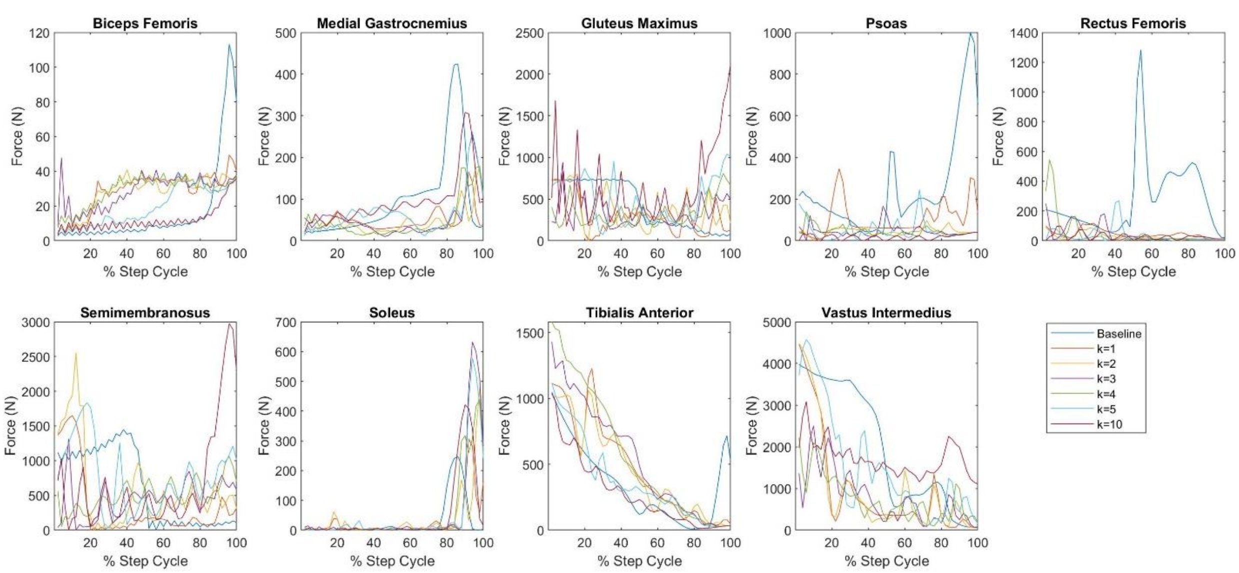

Figure 8: Muscle forces of hip spring assistance compared to baseline model

Overall, maximum muscle forces decreased for the biceps femoris when all joints were assisted. Maximum muscle forces for the vastus intermedius increased for both the ankle and knee springs. Maximum muscle forces decreased for the psoas for all knee and hip springs, and for the medial gastrocnemius and rectus femoris for all ankle and hip springs. All other maximum muscle forces varied based on spring location and stiffness.

Table 9: Summary of change in maximum muscle forces compared to baseline model for all joints and all spring stiffnesses

Joint	Increased Maximum Force	Decreased Maximum Force
Ankle	Tibialis Anterior, Vastus Intermedius	Biceps Femoris, Medial Gastrocnemius, Rectus Femoris
Knee	Vastus Intermedius	Biceps Femoris, Psoas
Hip	Soleus	Biceps Femoris, Medial Gastrocnemius, Psoas, Rectus Femoris

Maximum Forces Compared to Weakened Model

When the ankle was assisted, maximum muscle forces for the psoas, semimembranosus, soleus, and tibialis anterior increased, while no maximum muscle forces decreased (Table 9). Change in maximum muscle forces varied based on stiffness for the biceps femoris, medial gastrocnemius, gluteus maximus, rectus femoris, and vastus intermedius.

Table 10: Percent change in maximum muscle force of ankle assistance compared to weakened model

Stiffness (Nm/deg)	1	2	3	4	5	10
Biceps Femoris (Short Head)	2.30	0.20	1.86	2.33	1.54	-0.33
Medial Gastrocnemius	43.81	-3.49	-2.84	0.96	1.36	-0.58
Gluteus Maximus	-1.15	1.93	10.39	1.07	1.12	-0.64
Psoas	28.92	14.32	13.94	5.13	4.64	0.39
Rectus Femoris	-13.28	4.47	4.20	-1.64	-1.50	-0.17
Semimembranosus	22.58	7.22	7.13	34.34	33.85	1.30
Soleus	220.50	61.45	63.14	85.11	86.29	6.20
Tibialis Anterior	21.95	9.10	9.26	13.14	13.14	0.00
Vastus Intermedius	-2.35	5.20	5.23	5.77	5.78	-0.47

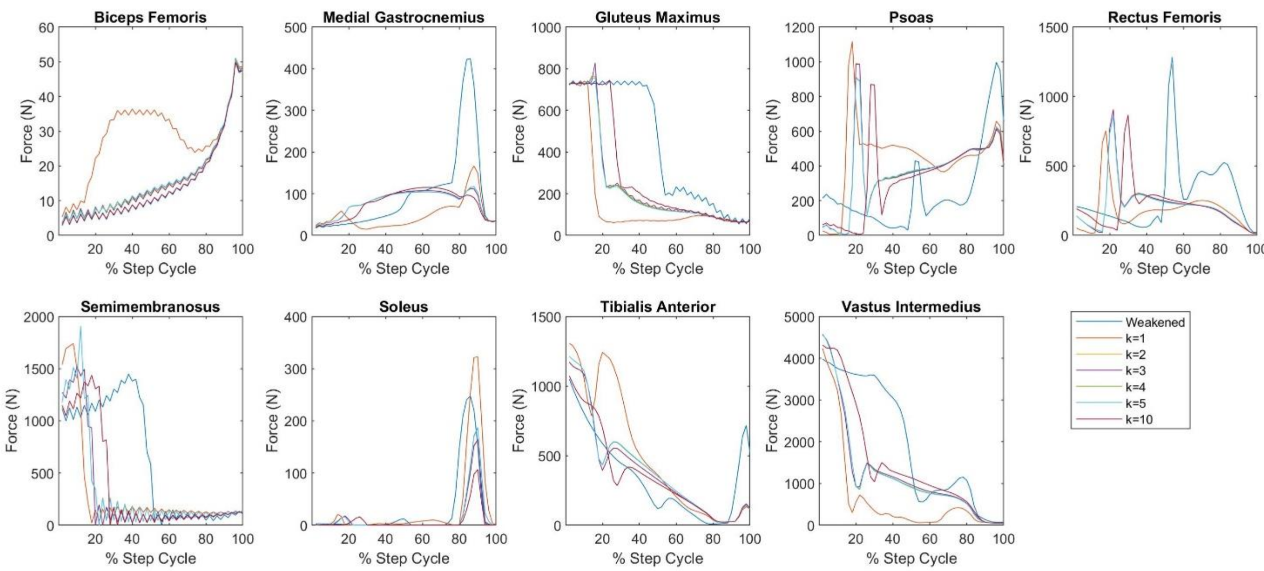

Figure 9: Muscle forces of ankle spring assistance compared to weakened model

When the knee was assisted, maximum muscle forces increased for the medial gastrocnemius, while maximum muscle forces decreased for the biceps femoris (Table 10). Change in maximum muscle force varied for the gluteus maximus, psoas, rectus femoris, semimembranosus, soleus, tibialis anterior, and vastus intermedius based on spring stiffness.

Table 11: Percent change in maximum muscle force of knee assistance compared to weakened model

Stiffness (Nm/deg)	1	2	3	4	5	10
Biceps Femoris (Short Head)	-30.06	-29.64	-29.50	-29.37	-29.22	-28.99
Medial Gastrocnemius	2.87	3.83	19.03	51.77	86.06	956.04
Gluteus Maximus	-1.97	0.91	14.39	9.08	5.41	18.10
Psoas	-10.34	-4.86	5.38	6.86	1.41	-39.86
Rectus Femoris	-2.15	-2.48	7.56	8.52	7.13	68.48
Semimembranosus	-4.28	17.14	41.20	-0.80	-2.51	-65.24
Soleus	-73.56	-37.73	5.97	67.45	139.65	1889.99
Tibialis Anterior	-4.32	12.12	22.21	5.94	-0.45	40.41
Vastus Intermedius	-2.79	4.90	3.61	4.29	0.53	36.33

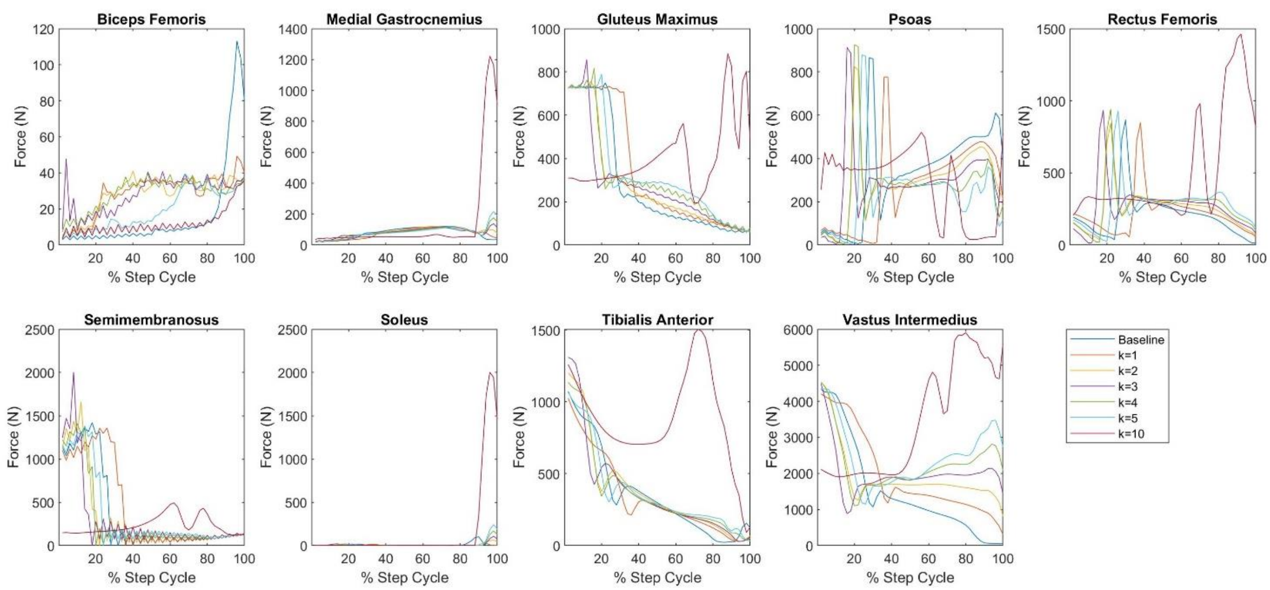

Figure 10: Muscle forces of knee spring assistance compared to weakened model

When the hip was assisted, maximum muscle forces for the biceps femoris, psoas, and rectus femoris decreased, while maximum muscle forces for the medial gastrocnemius and soleus increased significantly (Table 11). All other changes in maximum muscle forces varied with spring stiffness.

Table 12: Percent change in maximum muscle force of hip assistance compared to weakened model

Stiffness (Nm/deg)	1	2	3	4	5	10
Biceps Femoris (Short Head)	-1.28	-18.13	-4.20	-20.76	-20.76	-28.11
Medial Gastrocnemius	43.13	56.27	120.88	52.91	125.15	166.01
Gluteus Maximus	-2.33	7.11	25.56	7.76	38.63	182.81
Psoas	-60.15	-88.81	-80.85	-83.88	-71.67	-95.44
Rectus Femoris	-86.41	-89.77	-71.32	-37.35	-69.11	-88.54
Semimembranosus	16.18	80.06	-7.37	-25.23	29.27	109.64
Soleus	204.50	375.22	528.75	336.45	475.09	318.39
Tibialis Anterior	14.50	-1.00	33.41	47.50	4.31	-3.25
Vastus Intermedius	3.16	3.31	-42.13	-48.65	5.64	-28.73

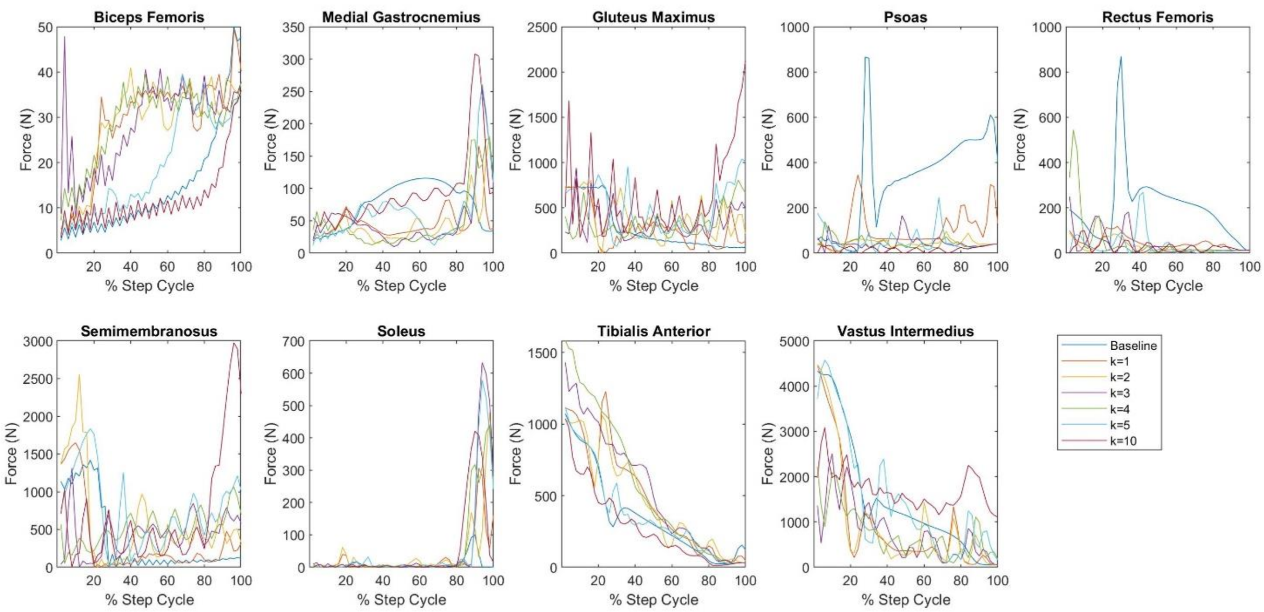

Figure 11: Muscle forces of hip spring assistance compared to weakened model

Overall, there were no muscles that had increased or decreased maximum muscle forces for all joint assistance at all stiffnesses.

Table 13: Summary of change in maximum muscle forces compared to weakened model for all joints and all spring stiffnesses

Joint	Increased Maximum Force	Decreased Maximum Force
Ankle	Psoas, Semimembranosus, Soleus, Tibialis Anterior	None
Knee	Medial Gastrocnemius	Biceps Femoris
Hip	Medial Gastrocnemius, Soleus	Biceps Femoris, Psoas, Rectus Femoris

3.3 Joint Kinematics

Kinematics with Ankle Assistance

When the ankle was assisted, all springs increased ankle angle from the weakened model in a similar way, closer to that of the baseline model (Figure 12). Similarly, the knee angle over the step cycle was shifted from the weakened model closer to the baseline model in a uniform way for all spring stiffnesses. Finally, the hip angles stayed relatively the same as the weakened and baseline models, while the spring of stiffness k =1 Nm/deg shifted the curve down and the spring with the stiffness k = 10 Nm/deg shifted the curve up and out.

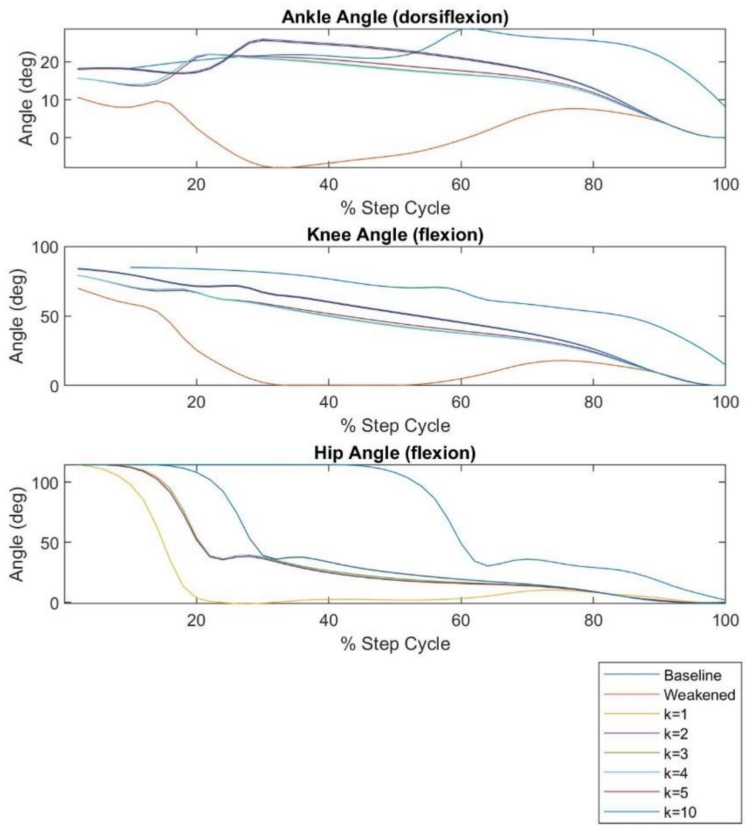

Figure 12: Joint angle response to ankle assistance

Kinematics with Knee Assistance

When the knee was assisted, the ankle angle shifted slightly, but stayed reltively the same for all spring stifflesses (Figure 13). Similarly, the knee angle stayed fairly consistent across springs of all siffnesses. Almost all spring stiffnesses shifted the hip angles back from the baseline and weakened models, while the spring with stiffness of 5 Nm/deg created a different hip angle curve.

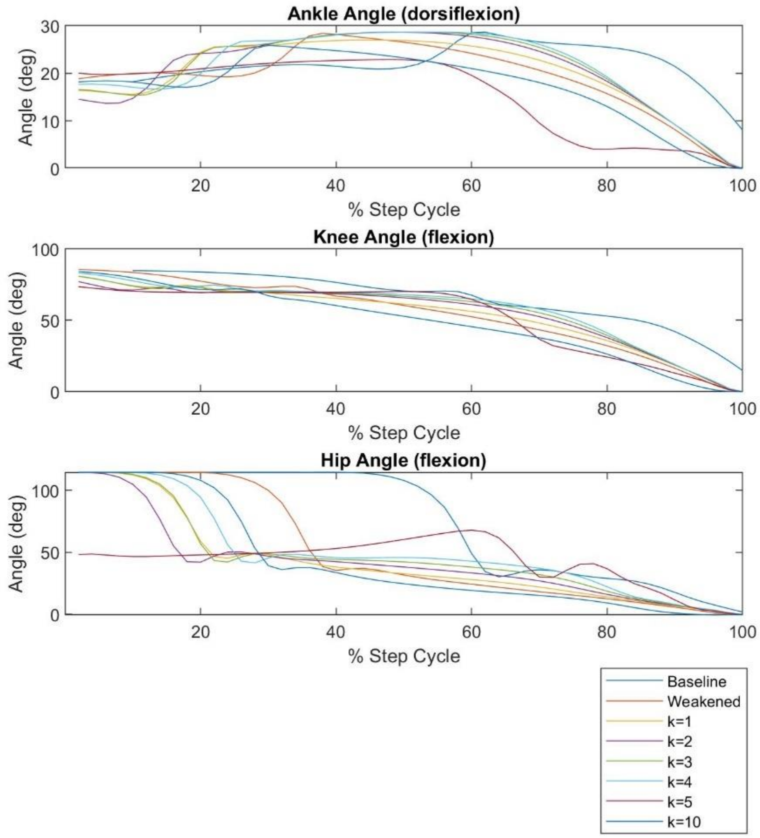

Figure 13: Joint angle response to knee assistance

Kinematics with Hip Assistance

When the hip was assisted, there was much more variation of joint angles (Figure 14). All springs shifted the ankle curve below the baseline model. Springs with stiffnesses 1 to 3 Nm/deg shifted the ankle angle curve below the weakened model, while stiffnesses 4 to 10 Nm/deg placed the curve between the weakened and baseline curves. Hip springs of all stiffnesses produced similar results at the knee as the ankle. Stiffnesses 1 to 3 Nm/deg shifted the curve below the weakened model, while stiffnesses 4 to 10 Nm/deg shifted the curve between the baseline and weakened models. For the hip angle, spring stiffnesses 1, 4, and 10 Nm/deg placed the curve between the weakened and baseline models. Stiffnesses 2, 3, and 5 Nm/deg created oscillation in the hip angle, consistent around 50 degrees. This oscillation and discrepancy between consecutive stiffnesses may be an error due to the simulation technology.

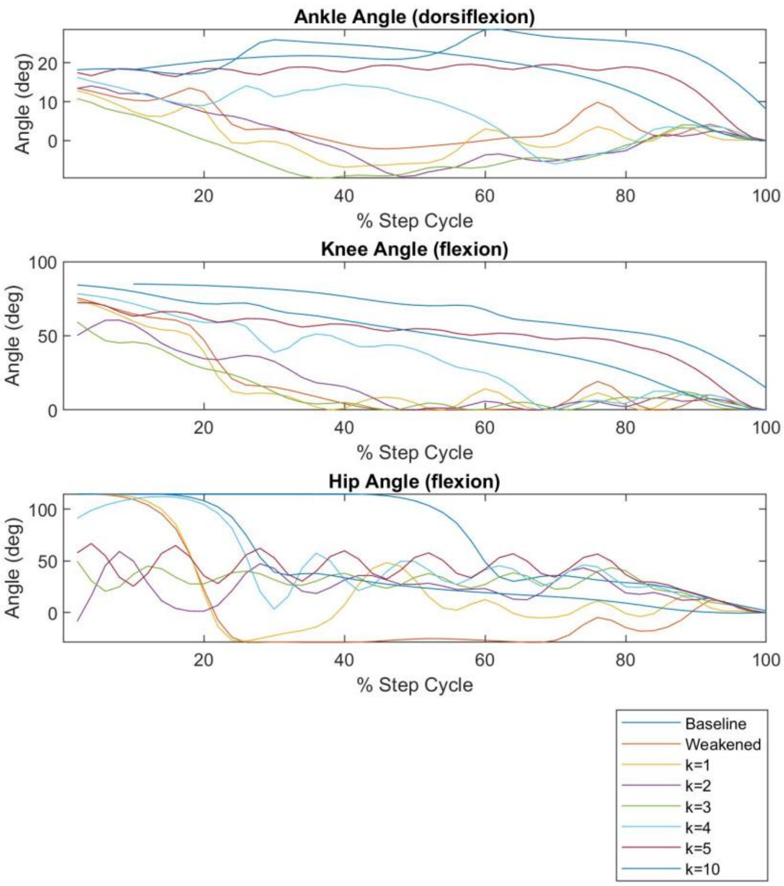

Figure 14: Joint angle response to hip assistance

4. Discussion and Conclusion

4.1 General Conclusions

All ankle springs with stiffness 1 to 5 Nm/deg slightly decreased overall metabolic cost. Since there was a weld joint between the foot and floor, there was not a large available range of ankle motion, which could explain why the changes are all less than 2%. Though five of the six springs decreased overall metabolic cost, all ankle springs increased more individual muscle costs than they decreased. Additionally, all ankle springs did not consistently decrease any maximum muscle forces compared to the weakened model.

The knee springs were the only springs that produced changes in metabolic cost that increased with increasing stiffness. All knee springs increased overall metabolic cost, ranging from 4.10% to 194.22%. The knee spring with stiffness k = 10 Nm/deg produced the worst results in regards to increasing metabolic costs and maximum muscle forces. Similar to the ankle, all knee springs consistently increased more muscle metabolic costs than they decreased. Compared to both the baseline and weakened models, there was not much variation in maximum muscle forces with spring assistance.

Across all metabolic costs and maximum muscle forces for hip assistance, there was variation in change based on spring stiffness. Overall metabolic cost varied in a nonuniform way, and individual muscle metabolic cost increased for three muscles and decreased for two. Compared to both the baseline and weakened models, the hip springs decreased maximum muscle forces more than they increased.

When the ankle was assisted, most joint angle curves were shifted from the weakened model closer to the baseline model for all joints. When the knee was assisted, the ankle and knee angles remained relatively unaffected, while the hip angle shifted below the baseline and weakened models. When the hip was assisted, there was much more variation based on stiffness and joint. Overall, the ankle springs restored joint kinematics closer to those of the baseline the most successfully out of all spring placements.

4.2 Limitations

Motion data was used from an older healthy individual, then max isometric forces were weakened to represent an individual with knee OA due to data accessibility issues. Using data from an individual with knee OA would produce more accurate results in responses to spring assistance. Additionally, a simplified, one leg model was used for the predictive simulation. This simulation replicated one step based on initial joint angles found from the muscle driven marker tracking simulation. A complex model with more degrees of freedom would produce different forces and metabolic costs since it would have more muscles to share force. Using the results from the MDMT simulation as input for the predictive simulation would provide more congruency between the tracking and predictive simulations, but this simulation was ultimately unsuccessful and could not be used. The springs used in the assistance were ideal and massless. An actual device designed with these springs would have a mass that would increase overall and muscle metabolic cost. This mass could be accounted for by adding it to the spring in future simulations.

4.3 Contributions

To our knowledge, this is the first study done to explore passive assistance for individuals with knee OA during stair ascent. Additional work should be done to provide more guidance of parameters for physical design of an assistive device.

4.4 Future Work

Next steps could be to rerun similar predictive simulations with more complex models and degrees of freedom to verify that these results are representative of muscle reactions to spring assistance at the lower limb joints. Additionally, running predictive simulations based off of knee OA kinematic data would provide more accuracy to reflect the knee OA population's compensated biomechanics [6,7,8]. Next steps could also be taken to use kinematic data from multiple individuals with knee OA to ensure that the device replicates similar results across different individuals.

Future work could also explore predictive simulations using different types of springs or combinations of springs. Adding a linear spring individually or in combination with the torsional spring could provide different insights than what was found in this study.

Summary

This thesis explored the effects of adding torsional springs with various stiffnesses to a weakened model during stair ascent. Results showed that all springs provided varying results based on stiffness and placement. No placement or stiffness had a positive impact on both metabolic cost and maximum muscle force.

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