Simulated Development of Assistive Devices to Aid Older Adults in Ascending Stairs

Undergraduate Honors Thesis

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By

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Abstract

Stair climbing is an important part of daily life. However, for older adults, stair climbing is one of the top five most difficult tasks, and the inability to climb stairs leads to a decreased quality of life. Assistive devices provide a way for people who cannot climb stairs to regain their mobility and improve their lives. While there are several assistive devices for climbing stairs on the market, assistive devices that use inexpensive elements like springs and assist joints like the knee and ankle have not been investigated. Simulations allow us to understand how assistive devices affect muscles during stair climbing and to test several variations of assistive devices before creating physical prototypes.

In this study, I used OpenSim, software that models the human musculoskeletal system, to add ideal, massless torsional springs to simulations of individuals ascending stairs. Four healthy participants (4 female, age = 65.00 ± 4.76 years, height = 1.61 ± 0.02 m, weight = 58.59 ± 6.11 kg) provided IRB-approved written consent. Motion capture and electromyography data were previously collected and used to create individual models in OpenSim. Static Optimization (SO) was used to resolve the kinematics of the individuals into forces and activations. Metabolic cost was estimated from the SO activations and compared to an individual with no assistance. In addition, maximum forces produced by certain muscles while ascending stairs were compared with and without varying assistive devices.

Overall metabolic cost increased for all spring stifffnesses and locations. The simulation of the unassisted individuals was the least metabolically expensive on average. However, two individuals had a decrease in overall metabolic cost when assisted at the ankle with a k =1 Nm/deg spring at the ankle, and one individual saw a decrease in metabolic cost when a

k = 1 Nm/ deg spring was located at the hip. The vastus lateralis, vastus intermedius, vastus medialis, gluteus maximus, and soleus decreased in metabolic cost for all spring stiffnesses and for all joints. Overall, a spring with stiffness k = 1 Nm/deg located at the ankle was the least metabolically expensive spring simulated in this study, increasing the cost by 3 ± 11%. A spring with stiffness k = 5 Nm/deg located at the knee was the most metabolically expensive device, increasing overall cost by 1421 ± 421%. The results of this study can be used to further develop assistive devices to help older adults climb stairs and ultimately improve their quality of life.

Acknowledgements

I would like to thank my advisor, Dr. Robert Siston. He helped encourage me to pursue undergraduate research and provided guidance and support this past year. I would not be where I am today or be going to graduate school next year if it were not for his encouragement and support. I am thankful for the opportunity to work in NMBL and find an area of mechanical engineering that makes me excited to continue my education.

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Chapter 1: Introduction

1.1 Background

Older Adults and Ascending Stairs

According to the Center for Disease Control, one in eight adults reported a mobility disability in 2013 [1]. A total of 30.6 million people reported that they had difficulty walking or climbing stairs in the 2010 census [2]. As people age, neurological and physiological capabilities decrease, like less muscle mass and muscle power [3]. Reduced muscular capabilities cause older adults to have difficulty ascending stairs, and as a result, decreased quality of life. Lower body weakness can cause consequences while ascending stairs such as broken bones and head injuries from falling [4]. In 2015, the total medical costs for falls were more than $50 billion [4]. Overall, limited mobility leads to a decreased quality of life, lack of independence, and decreased health for older adults [5].

Difficulties While Ascending Stairs and Capabilities

Ascending stairs becomes more difficult as people age and muscle strength decreases. Compared to younger people, older adults generated a lower maximum joint moment at the knee and ankle by 46% and 21% respectively while ascending stairs [6]. In a study of healthy elderly women climbing stairs, the main contributor to successful stair climbing was muscle power generated in the knee and ankle [7]. Outside factors such as gender, medication, and cognitive status had little effect on stair climbing capability [8].

Stair Climbing Compensation Strategies

Older adults use up to 93% of their maximum ankle joint moment while ascending stairs [9]. Studies show that people who cannot generate a maximum joint ankle moment of 1.5 Nm-kg may have difficulty ascending stairs without assistance [9]. Therefore, older adults have developed different strategies to compensate for decreased stair ascending capabilities. Compared to younger adults, older adults climbed stairs differently such as transferring energy from the knee to the ankle to increase the plantar flexion moment by using the gastrocnemius to reduce angular velocity at the knee [6]. Older adults also delayed the extension of the knee to create a larger knee joint moment and keep the ankle in dorsiflexion longer [6]. In addition to altering muscular strategies, older adults used handrails to prevent falling [9]. However, using a handrail only increased older adults' perception of stability, not their measured stability while ascending stairs. [9].

Current Assistive Devices

There are currently several assistive devices to aid older adults in ascending stairs (Figure 1).

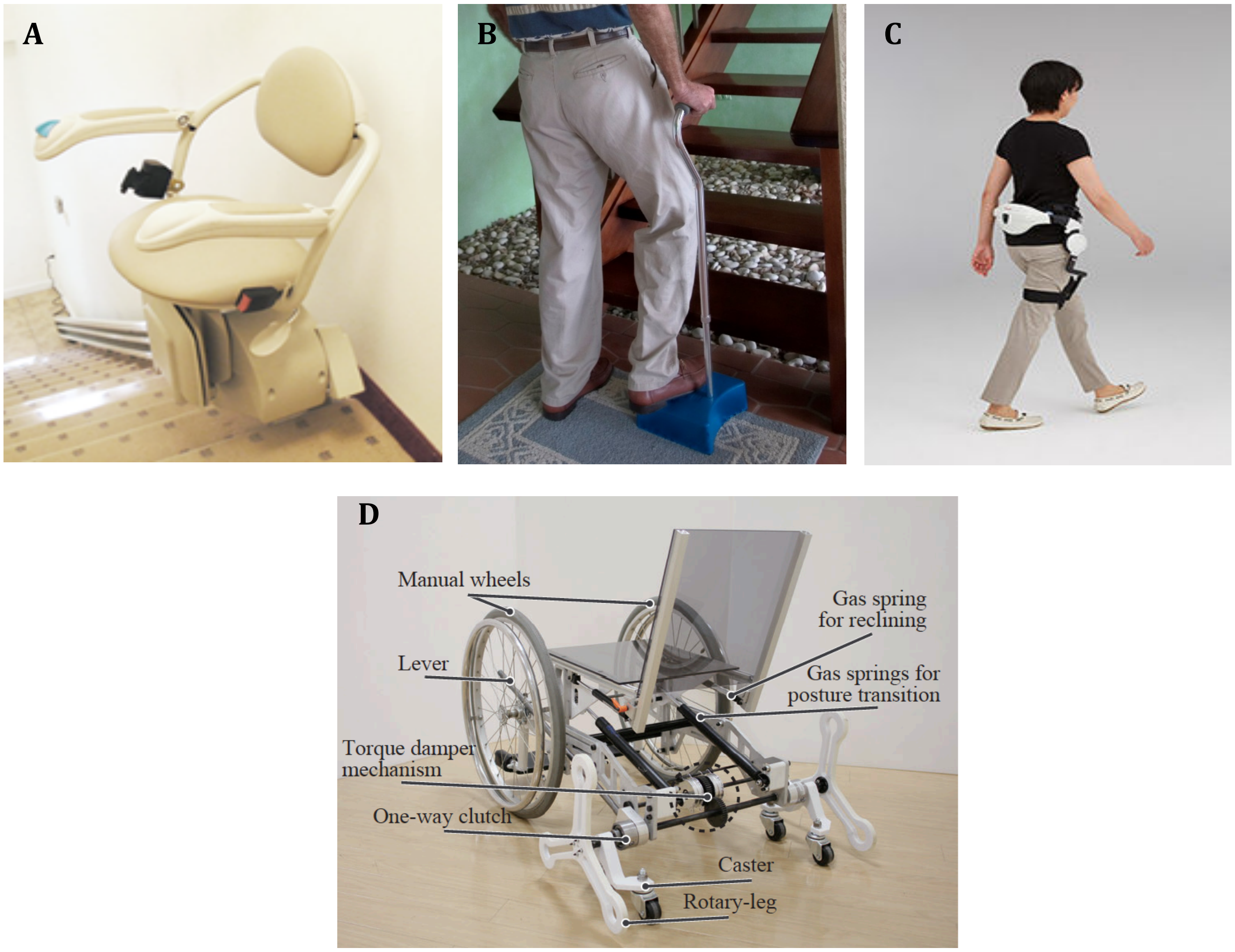

Figure 1: Current Assistive Devices

These devices have several disadvantages. The Stair Lift (Figure 1A) can be installed on a set of stairs and does not require the user to climb stairs. However, the high cost of the Stair Lift ranges from $1,800 to $5,800, not including the additional cost and time to install the device [10]. In addition, the Stair Lift lacks portability and must be installed on multiple sets of stairs. The EZ-STEP (Figure 1B) is an assistive device that is a half-step attached to a cane. Although this can make climbing easier by reducing hip and knee range of motion during stair ascent, the device weighs approximately two pounds and must be carried by the user [11]. The Honda

Walking Assist (Figure 1C) uses motors at the hip to assist with walking. The device weighs approximately 5.95 pounds and must be recharged every sixty minutes [12]. The heavy weight and the frequency of charging the battery make this device not feasible to wear for long periods of time. The Wheelchair with Lever Propulsion (Figure 1D) is an assistive device in development to allow people use arm strength and a one-way clutch to propel the wheelchair up stairs. The wheelchair requires arm strength, which all users may not have, and the user must reassemble the one-way clutch to go down stairs [13]. While there are current devices to assist elderly people in stair climbing, inexpensive and lightweight devices like springs and assistive devices at joints such as the knee and ankle have not been investigated for ascending stairs.

Dynamic Simulations of Movement

Dynamic simulations of human movement can be used to determine qualities that cannot be determined experimentally and can be used to better understand how assistive devices affect people's muscle forces and activations. OpenSim is an open source modeling and simulation software package that is used to study human movement [14], and past studies have used OpenSim to model tasks like gait, stair climbing, and the sit-to-stand transfer [15-18].

Simulations in OpenSim can answer several "what-if" questions, such as what happens when ideal actuators are virtually added to a musculoskeletal model. Changes in muscle recruitment such as muscle forces and activations as a result of adding these ideal actuators then can be examined. For example, by assisting a certain joint, certain muscles can produce less force [19]. By using simulations, several variations of assistive devices can be tested on multiple subjects before prototyping.

Simulating assistive devices can be an initial step in investigating how these devices impact muscle recruitment. Previous simulations of assistive devices have been applied to motions such as running, walking, and long jumping [19-21]. When active actuators were added to the ankle, knee, and hip, the length of the long jump increased from 2.27 m to 3.10 m, and the passive design increased the jump to 3.32 m [20]. When an assistive device was added to the ankle on individuals running at 2 m/s, the average metabolic cost decreased 26 % compared to unassisted running [19]. When a passive assistive device was added to the hip, the metabolic cost of walking decreased by up to 10% during simulation [21]. A passive assistive device for ascending stairs could potentially help people with muscle weakness successfully climb stairs.

1.2 Focus of Thesis

The focus of this research was to investigate the effects of simulated assistive devices on the muscle forces and metabolic cost of older adults ascending stairs.

1.3 Significance of Research

There are several assistive devices currently on the market to aid people in climbing stairs. However, there are few, if any, devices that are inexpensive and simple to use. The use of assistive devices at joints like the knee and ankle has not been investigated. Simulating assistive devices based on the motion of older adults will help developers better understand how to assist people in ascending stairs, prototype, and eventually test assistive devices.

1.4 Overview of Thesis

This thesis contains five chapters. The second chapter presents the methods of the simulations of the assistive device. The third chapter presents the results of the simulations of the assistive devices during the stair ascent cycle. The fourth chapter analyzes and discusses the results of the simulations from this study. The fifth chapter presents the conclusions from these simulations.

Chapter 2: Methods

2.1 Experimental Data

Four subjects (4 female, age = 65.00 ± 4.76 years, height = 1.61 ± 0.02 m, weight = 58.59 ± 6.11 kg) provided written informed consent in accordance with the Institutional Review Board of The Ohio State University. All patients had no known lower limb or nervous system pathologies.

Stair Ascent Motion Data

Experimental motion data of stair ascent were collected by Dr. Elena Caruthers and Dr. Sarah Roelker in the Motion Analysis and Performance Lab (The Ohio State University). Subjects ascended a custom staircase (tread depth: 25.5 cm, step height: 20 cm) at a self-selected speed.

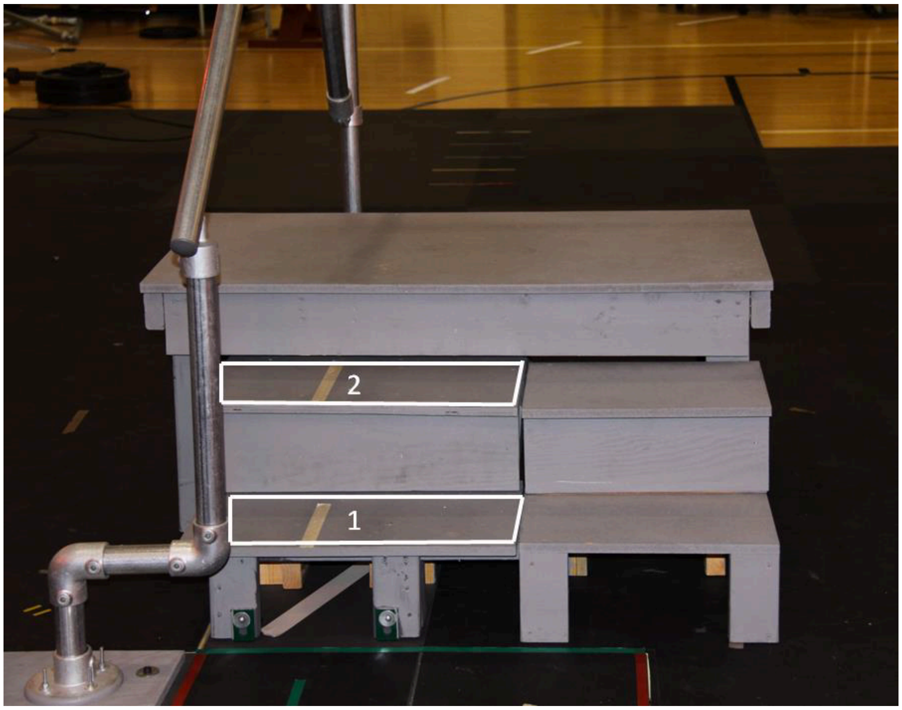

Figure 2: Custom staircase with ground reaction force plates

Motion data were collected at 150 Hz using a 10-camera Vicon MX-F40 system (Centennial, CO), and the Modified Point Cluster Technique with additional markers on the iliac crests were used to track the lower limbs [22]. Ground reaction forces were measured through three force plates in the floor and below the first two steps. Force plate data were sampled at 600 Hz (Bertec, Columbus, OH). Bilateral, 16-channel, surface electromyography data were collected from the rectus femoris, vastus medialis, vastus lateralis, medial gastrocnemius, lateral gastrocnemius, soleus, semimembranosus, and biceps femoris and sampled at 1500 Hz (Telemyo DTS, Noraxon USA, Inc; Scottsdale, AZ). EMG data were high-passed filtered at 10 Hz, rectified, and RMS smoothed with a 20 ms window.

Subjects ascended stairs at a self-selected speed for six trials, and one trial was selected to be used for simulation. Subjects also performed a hip joint center calibration, and the hip joint center trials were analyzed to add a hip joint center marker on the Vicon trials [23]. C3D Extraction Toolbox was used to extract experimental marker positions and EMG data from Vicon to a format compatible with OpenSim via MATLAB [24].

2.2 Musculoskeletal Modeling

Subject specific scaled models were created from the Full Body Model 2016 containing 94 muscles, 46 degrees of freedom, and a flexible spine [14]. Anne Marie Jackson, a former graduate student, used OpenSim 3.3 to run Inverse Kinematics, Residual Reduction Algorithm, Inverse Dynamics, and Static Optimization (Figure 3) [14].

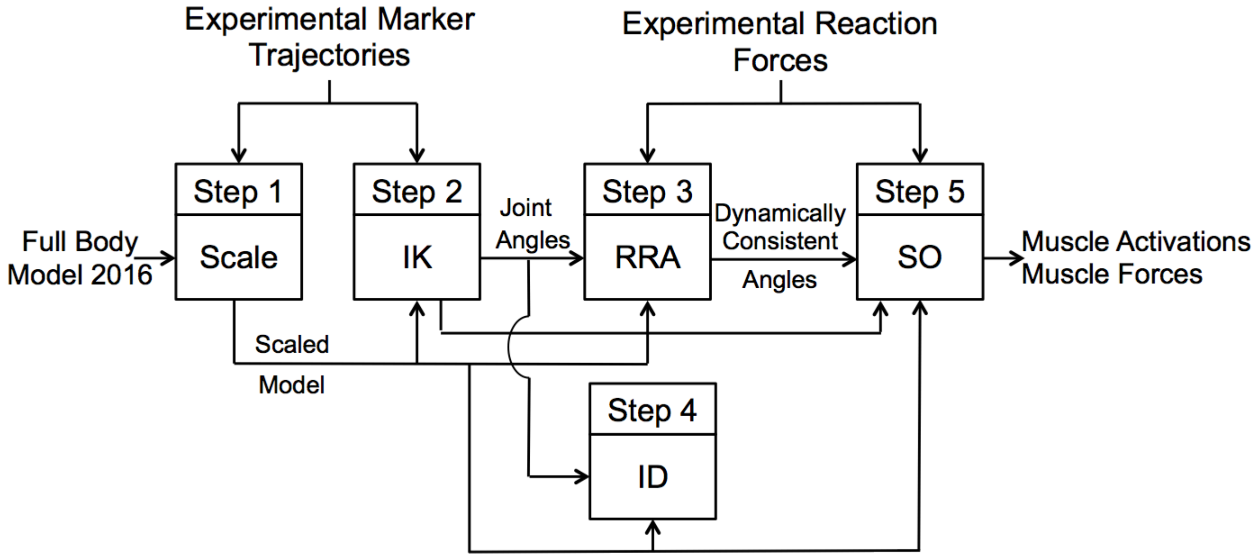

Figure 3: OpenSim process

Dimensional Scaling of Models

Models were scaled to match experimental makers in the static calibration trial. Anne Marie adjusted masses of each models to equal the masses of individuals, while the mass distribution among segments matched the original model. Additionally, relative weights of various virtual and experimental markers were adjusted until the RMS marker error was less than 3 cm [25].

Inverse Kinematics

Anne Marie solved Inverse Kinematics by minimizing the difference between experimental and virtual markers. She used a least-squares approach throughout the stair ascent cycle, and she adjusted the marker weights for different time points to reduce RMS error to below 3 cm [25].

Residual Reduction Algorithm

Anne Marie used the Residual Reduction Algorithm (RRA) to reduce the dynamic inconsistencies between the model kinematics calculated from Inverse Kinematics and the experimental ground reaction forces. Modeling assumptions and experimental error created the dynamic inconsistencies between simulations and experimental data. RRA re-distributed the masses among segments, and changed the center of the mass of the torso to counteract experimental errors and dynamic inconsistencies. OpenSim best practices were used to determine when the residual forces and residual moments were acceptable [25]. RRA was run for only part of the stair ascent cycle, because in the experimental setup, the top step was not instrumented with a force plate. RRA requires ground reaction forces for both feet, so Inverse Kinematics data was used for portions of the data where there were only the ground reaction forces for one foot. After running RRA, Anne Marie combined kinematic data from RRA and IK using a spline function. There was then one continuous curve for the stair ascent cycle.

Inverse Dynamics

Anne Marie used the Inverse Dynamics (ID) tool to determine net joint torques for the parts of stair ascent where ground reaction forces were not collected [14]. The ID tool used double differentiation to estimate the accelerations from the experimental motion and ground reaction forces. One joint torque curve was created by splining the ID and RRA torque data.

Static Optimization

Static Optimization (SO) was used to estimate individual muscle forces from joint torques by minimizing the sum of muscle activations squared at each time frame of the trial [25]. Anne Marie used both RRA and IK kinematics to run SO. The SO results were splined together to analyze forces and activations for the entire stair ascent cycle.

Simulated and Experimental Data Agreement

Anne Marie compared simulated and experimental muscle activations to make sure the simulated models reflected the experimental data. The overall pattern and timing of the activations matched even though peak magnitudes differ (Figure 4). Anne Marie calculated RMS error, and error was below 0.3 for all muscles [26].

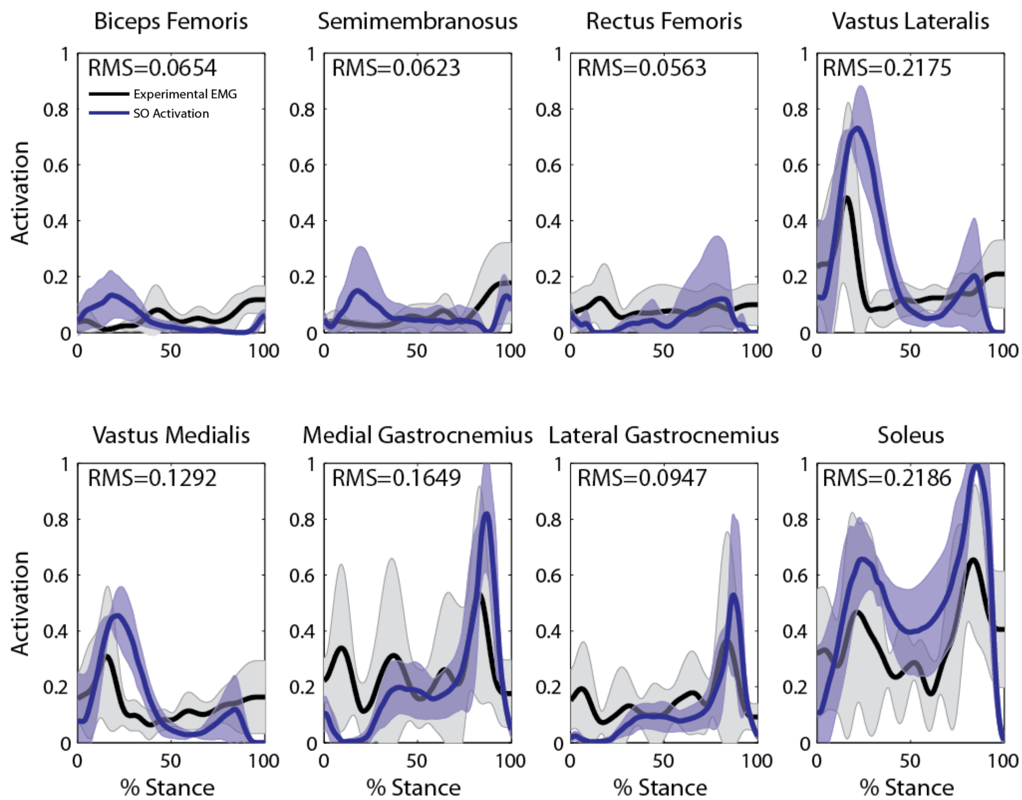

Figure 4: Plot of experimental EMG compared to SO activations for one representative individual [15]

Anne Marie compared joint torques from SO to the normalized joint torques calculated from RRA and ID. SO joint torques were calculated by summing the product of muscle forces and their corresponding moment arms throughout the stair ascent cycle. Anne Marie compared torques to check if the muscles were producing the joint torques and not the reserve actuators, additional joint torques that augment muscle forces.

2.3 Modeling Assistive Device

I, then, added one ideal, massless torsional spring at a time to the ankle, knee, and hip with varying stiffnesses from 1 to 5 Nm/deg in 1 Nm/deg increments. The torsional springs were modeled using the Coordinate Limit Force function and had a neutral angle of zero degrees. I ran sixty simulations, fifteen per subject. The spring at the ankle assisted in plantar flexion. The spring at the hip assisted in extension. The spring at the knee assisted in extension.

Repeat Static Optimization

I repeated Static Optimization for the assisted simulations with the experimental kinematics (Figure 5). RRA and IK kinematics were both used to run SO, and splined together.

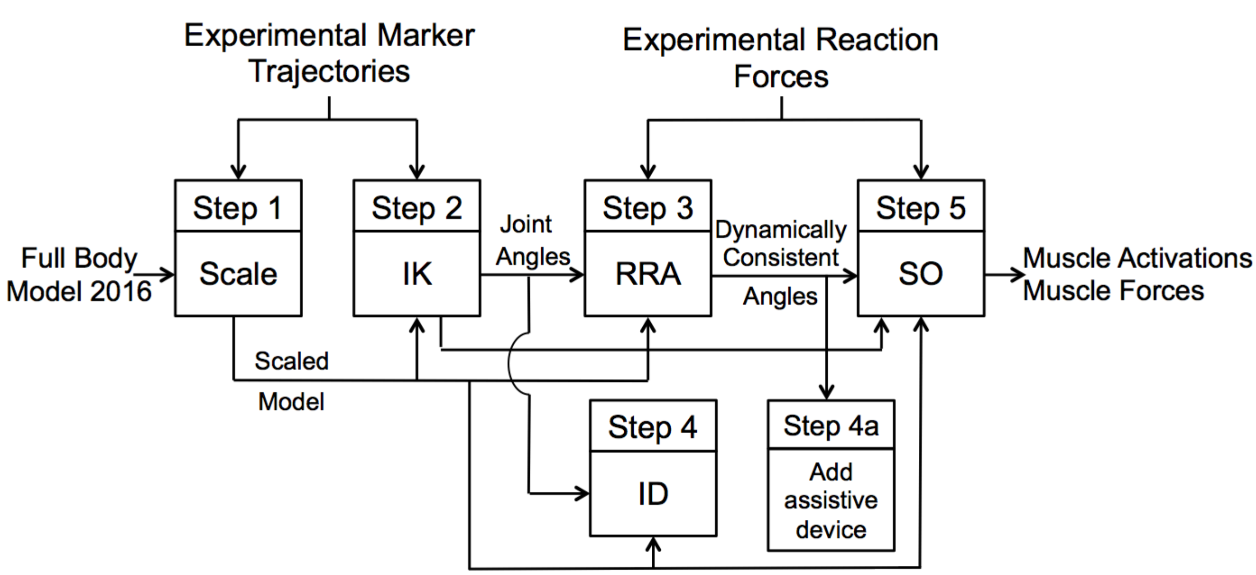

Figure 5: OpenSim process with added actuator

2.4 Analysis

Metabolic Cost

I estimated metabolic cost for the lower extremity muscles of the assisted leg. Metabolic cost was estimated by the sum of activations squared for each muscle at each time point (Equation 1) [27]. Metabolic costs for individual muscles over the entire stair ascent cycle were summed together to estimate the overall metabolic cost. Metabolic cost was then compared to a baseline of the unassisted individual.

$$Metabolic\ Cost = \sum Muscle\ Activation^2 \text{ (Equation 1)}$$

Forces

I also compared the maximum force from Static Optimization for each muscle in the affected leg. The maximum force was compared to the baseline of the model with no assistive device.

3. Results

Results are presented in the following order: overall metabolic cost, individual metabolic cost for muscles, and maximum muscle forces. Metabolic costs and maximum muscle forces are presented in percent changes as compared to an unassisted participant.

3.1 Cost

Overall Metabolic Cost

The average overall metabolic cost increased for all cases. However, overall metabolic cost decreased for two participants when the 1 Nm/deg spring was at the ankle and for one participant when the 1 Nm/deg spring was at the hip. Metabolic cost increased the least of all the springs when the participant was assisted at the ankle with a 1 Nm/deg spring (Table 1). Metabolic cost increased as spring stiffness increased. The most metabolically expensive device was the spring with stiffness 5 Nm/deg located at the knee.

Table 1: Average % change in metabolic cost compared to individuals with no assistance

Joint	Stiffness (Nm/deg)	Overall % Change
Ankle	1	3 ± 11%
	2	44 ± 38%
	3	90 ± 72%
	4	112± 84%
	5	138 ± 91%
Knee	1	157 ± 100%
	2	406 ± 110%
	3	976 ± 248%
	4	1290 ± 334%
	5	1421 ± 421%
Hip	1	15 ± 25%
	2	104 ± 67%
	3	201 ± 93%
	4	385 ± 164%
	5	544 ± 208%

Metabolic Cost for Muscle Groups

While metabolic cost increased overall, metabolic cost for certain muscles increased and decreased depending on which joint was assisted. When the ankle was assisted, metabolic cost increased in the iliacus, tibialis anterior, biceps femoris long head, semimembranosus, gluteus medius, gluteus minimus, and medial gastrocnemius for all spring stiffnesses (Table 2). However, metabolic cost decreased in the vatus lateralis, vastus intermedius, vastus medialis, gluteus maximus, and soleus for all spring stiffnesses. The metabolic cost of the tibialis anterior increased 30512% when assisted at the ankle with a 5 Nm/deg torsional spring because the metabolic cost of the tibialis anterior for one representative individual was 0.9 when unassisted, and the tibialis anterior metabolic cost was 514 when the spring was added.

Table 2: Average % change in metabolic cost when assisted at the ankle and compared to individuals with no assistance

Stiffness (Nm/deg)	1	2	3	4	5
Iliacus (Iliac)	604 ± 1047%	38 ± 109%	38 ± 108%	39 ± 108%	41 ± 108%
Rectus Femoris (RecFem)	-22 ± 46%	4 ± 23%	2 ± 23%	0 ± 22%	-1 ± 22%
Vastus Lateralis (VasLat)	-5 ± 3%	-8 ± 5%	-9 ± 6%	-10 ± 6%	-11 ± 7%
Vastus Intermedius (VasInt)	-5 ± 3%	-7 ± 5%	-9 ± 6%	-10 ± 6%	-10 ± 7%
Vastus Medialis (VasMed)	-5 ± 3%	-7 ± 4%	-9 ± 5%	-10 ± 6%	-11 ± 6%
Tibialis Anterior (TibAnt)	3240 ± 3489%	10882 ± 9452%	15842 ± 12248%	21657 ± 15823%	30512 ± 23871%
Gluteus Medius (GluteMed)	2 ± 4%	1 ± 1%	1 ± 1%	1 ± 1%	2 ± 1%
Gluteues Maximus (GluteMax)	-1 ± 0%	-2 ± 1%	-2 ± 1%	-2 ± 1%	-2 ± 1%
Gluteus Minimus (GluteMin)	7 ± 13%	1 ± 0%	2 ± 1%	2 ± 1%	3 ± 2%
Biceps Femoris Long Head (BiFemLH)	16 ± 29%	1 ± 2%	1 ± 2%	0 ± 2%	0 ± 2%
Semimembranosus (Semimem)	26 ± 15%	14 ± 12%	15 ± 9%	14 ± 11%	18 ± 22%
Lateral Gastrocnemius (GasLat)	-6 ± 18%	48 ± 112%	158 ± 333%	167 ± 339%	202 ± 330%
Medial Gastrocnemius (GasMed)	9 ± 29%	34 ± 66%	33 ± 73%	34 ± 67%	46 ± 62%
Soleus	-23 ±7%	-30 ± 11%	-38 ± 12%	-43 ± 12%	-47 ± 12%

When the knee was assisted, metabolic cost increased in the illiacus, tibialis anterior, biceps femoris long head, semimembranosus, lateral gastrocnemius, gluteus medius, gluteus minimus, medial gastrocnemius for all spring stiffnesses (Table 3). However, metabolic cost decreased for the gluteus maximus, vastus lateralis, vastus intermedius, vastus medialis, and soleus. The vastus intermedius and vastus medialis decreased to 100 % for spring stiffness k =2, 3, and 4 Nm/deg. The metabolic cost of the iliacus increased 86572% when assisted at the knee with a 5 Nm/deg torsional spring because the metabolic cost of the iliacus for one representative individual was 0.6 when unassisted, while when a spring was added the total iliacus metabolic cost was 1308. The metabolic cost of the semimembranosus increased 25664% when assisted at the knee with a 5 Nm/deg torsional spring because the metabolic cost of the semimembranosus for one representative individual was 6 when unassisted, while when a spring was added the semimembranosus metabolic cost was 1308.

Table 3: Average % change in metabolic cost when assisted at the knee compared to individuals with no assistance

Stiffness (Nm/deg)	1	2	3	4	5
Iliacus (Iliac)	16094 ± 8820%	52962 ± 23483%	115049 ± 70515%	137184 ± 86554%	137186 ± 86572%
Rectus Femoris (RecFem)	4750 ± 3549%	1609 ± 1341%	609 ± 897%	213 ± 370%	-100 ± 0%
Vastus Lateralis (VasLat)	-96 ± 3%	-99 ± 3%	-99 ± 3%	-99 ± 3%	-99 ± 3%
Vastus Intermedius (VasInt)	-97 ± 1%	-100 ± 0%	-100 ± 0%	-100 ± 0%	-100 ± 0%
Vastus Medialis (VasMed)	-96 ± 3%	-100 ± 0%	-100 ± 0%	-100 ± 0%	-100 ± 0%
Tibialis Anterior (TibAnt)	9242 ± 7070%	14698 ± 12299%	27420 ± 30826%	32396 ± 39914%	32398 ± 39916%
Gluteus Medius (GluteMed)	31 ± 10%	176 ± 61%	415 ± 147%	488 ± 160%	481 ± 179%
Gluteues Maximus (GluteMax)	-43 ± 13%	-76 ± 12%	-53 ± 43%	-71 ± 41%	-75 ± 42%
Gluteus Minimus (GluteMin)	47 ± 11%	503 ± 253%	7579 ± 3856%	12887 ± 3230%	15243 ± 3985%
Biceps Femoris Long Head (BiFemLH)	1249 ± 1043%	1448 ± 1503%	4860 ± 2990%	7161 ± 3518%	8699 ± 3798%
Semimembranosus (Semimem)	4462 ± 4389%	9105 ± 8001%	18016 ± 14056%	23894 ± 19182%	25664 ± 21872%
Lateral Gastrocnemius (GasLat)	193 ± 178%	1348 ± 1327%	2822 ± 2720%	3337 ± 3233%	3387 ± 3272%
Medial Gastrocnemius (GasMed)	371 ± 172%	880 ± 421%	1036 ± 555%	1051 ± 569%	1050 ± 569%
Soleus	-41 ±6%	-69 ± 13%	-72 ± 15%	-72 ± 15%	-72 ± 15%

When the spring was located at the hip, the metabolic cost increased in the iliacus, rectus femoris, tibialis anterior, lateral gastrocnemius, gluteus minimus, and medial gastrocnemius for all spring stiffnesses (Table 4). Metabolic cost decreased for the vastus lateralis, vastus intermedius, vastus medialis, gluteus medius, gluteus maximus, biceps femoris long head, semimembranosus, and soleus for all spring stiffnesses. The biceps femoris long head metabolic cost decreased 100%. The metabolic cost of the iliacus increased 132266% when assisted at the hip with a 5 Nm/deg torsional spring because the metabolic cost of the iliacus for one representative individual was 1239, while the metabolic cost of the iliacus was 0.6 when unassisted. The metabolic cost of the rectus femoris increased 79042% when assisted at the hip with a 5 Nm/deg torsional spring because the metabolic cost of the rectus femoris for one representative individual was 1004, while the metabolic cost of the rectus femoris was 1 when unassisted.

Table 4: Average % change in metabolic cost when assisted at the hip and compared to individuals with no assistance

Stiffness (Nm/deg)	1	2	3	4	5
Iliacus (Iliac)	6315 ± 4206%	30984 ± 19764%	73697 ± 46657%	112279 ± 70130%	132266 ± 83948%
Rectus Femoris (RecFem)	10779 ± 3026%	42512 ± 23490%	64727 ± 45065%	73721 ± 51724%	79042 ± 55384%
Vastus Lateralis (VasLat)	-51% ± 4%	-79% ± 3%	-83% ± 5%	-83% ± 6%	-83% ± 6%
Vastus Intermedius (VasInt)	-53 ± 5%	-81 ± 3%	-85 ± 7%	-84 ± 7%	-84 ± 7%
Vastus Medialis (VasMed)	-46 ± 13%	-76 ± 9%	-83 ± 5%	-82 ± 5%	-81 ± 5%
Tibialis Anterior (TibAnt)	1847 ± 1187%	4095 ± 2594%	4216 ± 2717%	4474 ± 2955%	5669 ± 4680%
Gluteus Medius (GluteMed)	-22 ± 4%	-30 ± 9%	-32 ± 14%	-33 ± 28%	-20 ± 42%
Gluteues Maximus (GluteMax)	-85 ± 7%	-99 ± 2%	-100 ± 0%	-100 ± 0%	-100 ± 0%
Gluteus Minimus (GluteMin)	1 ± 4%	63 ± 94%	420 ± 474%	3652 ± 2271%	7232 ± 4326%
Biceps Femoris Long Head (BiFemLH)	-100 ± 0%	-100 ± 0%	-100 ± 0%	-100 ± 0%	-100 ± 0%
Semimembranosus (Semimem)	-73 ± 24%	-88 ± 9%	-96 ± 3%	-99 ± 3%	-99 ± 2%
Lateral Gastrocnemius (GasLat)	26 ± 19%	74 ± 52%	142 ± 109%	201 ± 177%	281 ± 299%
Medial Gastrocnemius (GasMed)	46 ± 32%	124 ± 68%	223 ± 96%	305 ± 130%	362 ± 175%
Soleus	-8 ±1%	-22 ± 2%	-31 ± 4%	-35 ± 5%	-38 ± 7%

Overall, the metabolic cost increased the least of all the springs from the baseline of the unassisted individual when a 1 Nm/deg spring was located at the ankle, and metabolic cost increased the most when a 5 Nm/deg spring was located at the knee.

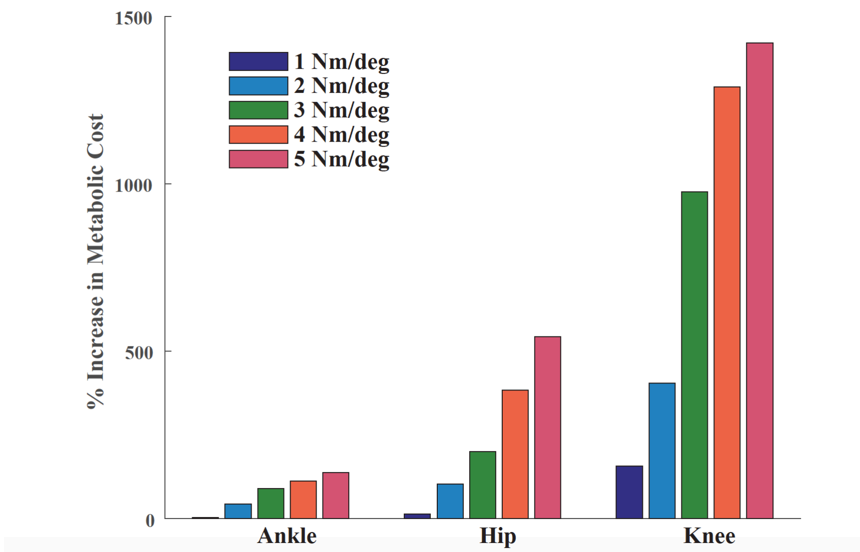

Figure 6: Average of the overall metabolic cost for individuals assisted at the ankle, knee, and hip with spring stiffnesses ranging from 1 Nm/deg to 5 Nm/deg

Overall, the iliacus, tibialis anterior, gluteus minimus, and medial gastrocnemius increased for spring stiffnesses and for all joints (Table 5). However, the vastus lateralis, vastus intermedius, vastus medialis, gluteus maximus, and soleus decreased in metabolic cost for all spring stiffnesses and for all joints.

Table 5: Summary of % change for all joints and all spring stiffnesses. Text in red represents muscles that increased or decreased in % cost respectively for all cases.

Joint	↑ % Cost	↓ % Cost
Ankle	Iliac, TibAnt, BiFemLH, Semimem, GasMed, GluteMed, GluteMin	VasLat, VasInt, VasMed, Soleus, GluteMax
Knee	Iliac, TibAnt, BiFemLH, Semimem, GasLat, GasMed, GluteMed, GluteMin	VasLat, VasInt, VasMed, Soleus, GluteMax
Hip	Iliac, RecFem, TibAnt, GasLat, GasMed, GluteMin	VasLat, VasInt, VasMed, BiFemLH, Semimem, Soleus, GluteMax, GluteMed

3.2 Muscle Activations

When the ankle was assisted, activations in the tibialis anterior increased as spring stiffness increased from 1 to 5 Nm/deg, while activations in the soleus, the muscle that assists in plantarflexion, decreased as spring stiffness increased (Figure 6). Activations in the vastus intermedius, vastus lateralis, vastus medialis, gluteus maximus, gluteus medius, and gluteus minimus stayed almost constant as spring stiffness increased.

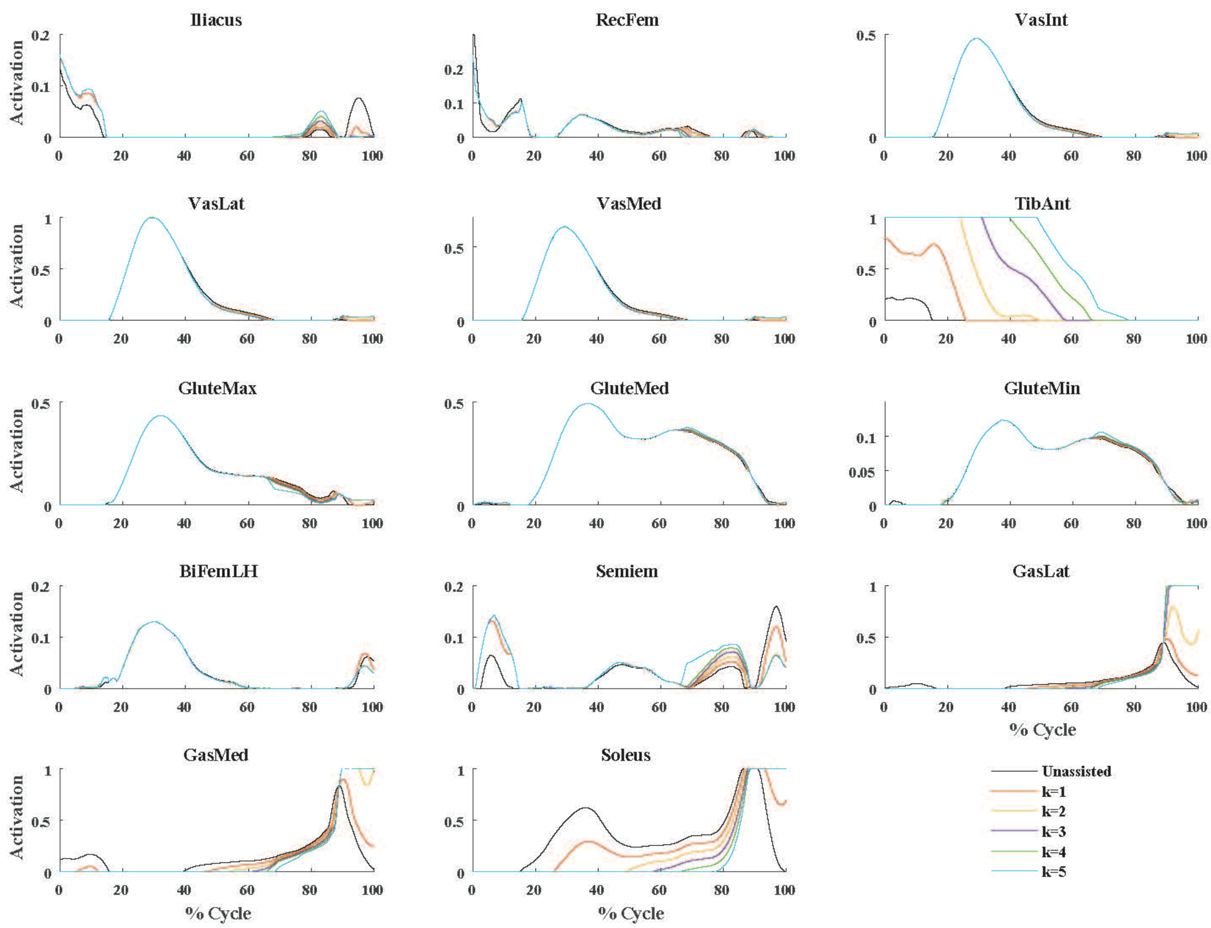

Figure 7: Selected muscle activations for one representative individual assisted at the ankle at varying spring stiffnesses in Nm/deg

When the knee was assisted, activations in the iliacus, tibialis anterior, gluteus minimus, semimembranosus, lateral gastrocnemius, and medial gastrocnemius increased as spring stiffness increased (Figure 7). The iliacus, tibialis anterior, gluteus minimus, semimbranosus, and lateral gastrocnemius, biceps femoris long head reached full activation during the stair ascent cycle. Activations in the vastus intermedius, vastus lateralis, and vastus medialis decreased as spring stiffness, and soleus decreased. The vastus intermedius, vastus lateralis, vastus medialis, and soleus had little to no activation during parts of the stair ascent cycle when the spring stiffness were k = 3, 4, and 5 Nm/deg.

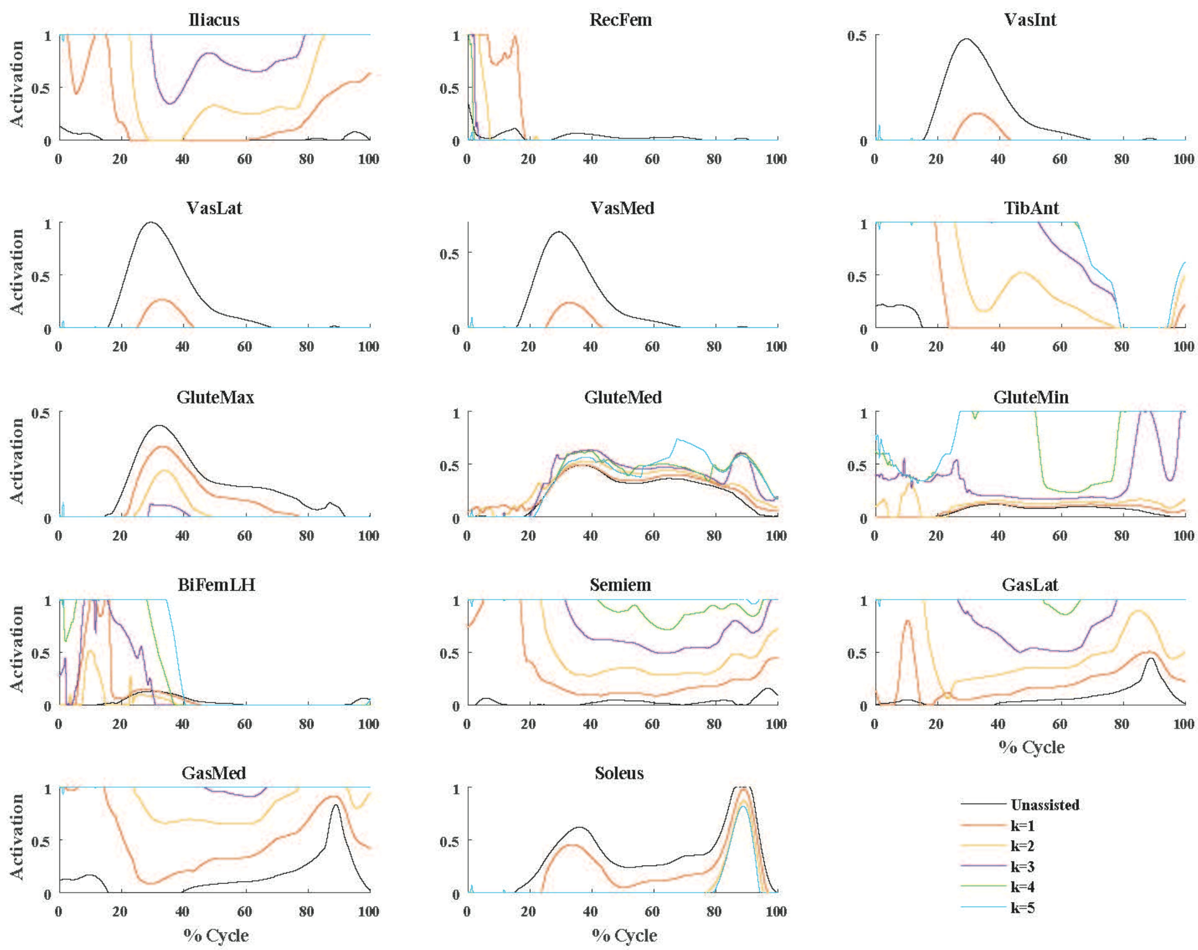

Figure 8: Selected muscle activations for one representative individual assisted at the knee at varying spring stiffnesses in Nm/deg

When the hip was assisted, activations in the iliacus, rectus femoris, tibialis anterior, gluteus minimus, lateral gastrocnemius, and medial gastrocnemius increased as spring stiffness increased (Figure 8). Activations in the vastus intermedius, vastus lateralis, vastus medialis, gluteus maximus, biceps femoris long head, gluteus medius, and semimembranosus decreased as spring stiffness increased.

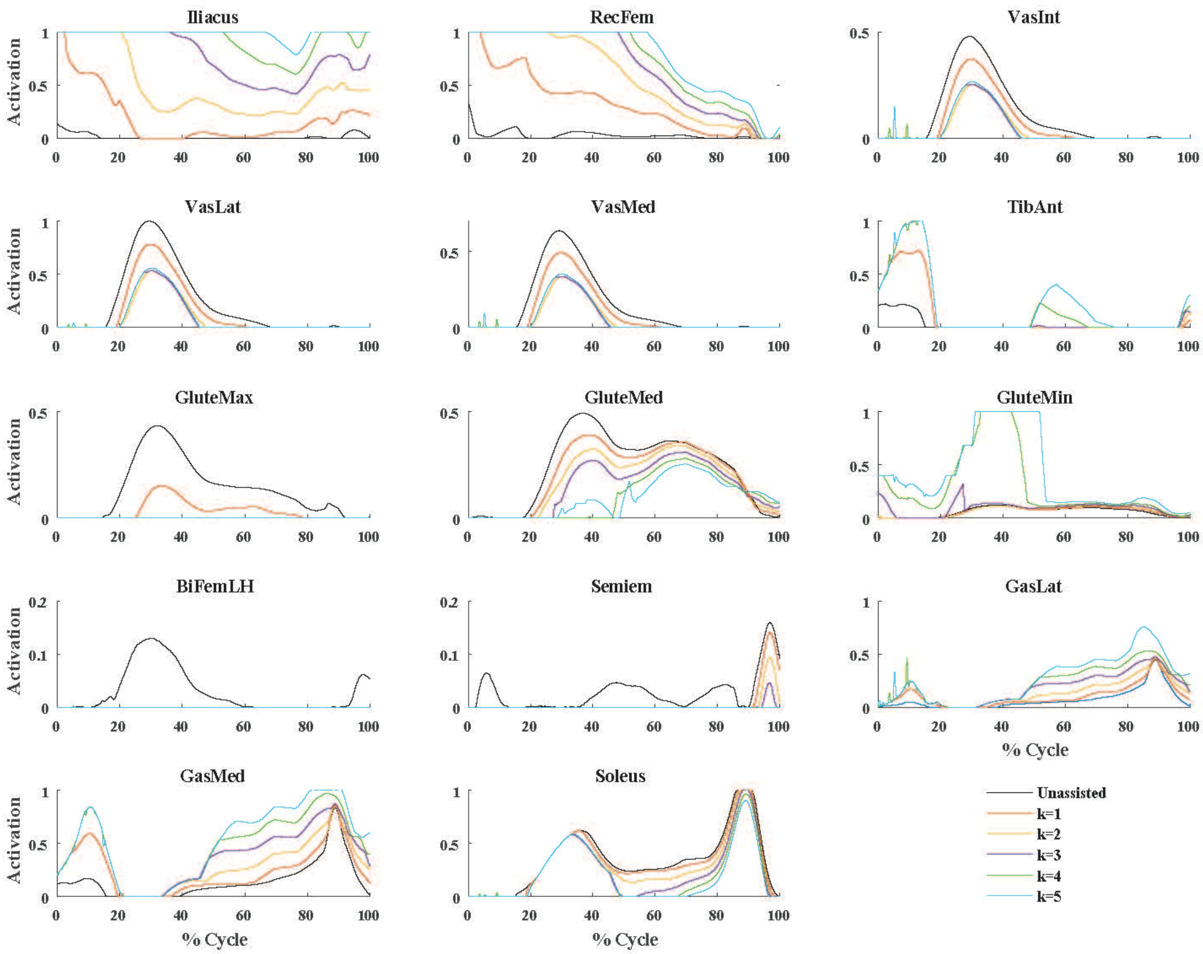

Figure 9: Selected muscle activations for one representative individual assisted at the hip at varying spring stiffnesses in Nm/deg

3.3 Muscle Forces

When the ankle was assisted, maximum forces in the iliacus, vastus lateralis, vastus intermedius, tibialis anterior, biceps femoris long head, and semimembranosus increased (Figure 9). However, maximum forces in the rectus femoris, lateral gastrocnemius, medial gastrocnemius, vastus medialis, and soleus decreased. Maximum force in the tibialis anterior increased more compared to other muscles (Table 6).

Table 6: Average % change in maximum force of selected muscles in individuals assisted at the ankle

Stiffness (Nm/deg)	1	2	3	4	5
Iliacus (Iliac)	125 ± 174%	60 ± 46%	60 ± 46%	60 ± 46%	60 ± 46%
Rectus Femoris (RecFem)	-12 ± 14%	-12 ± 15%	-12 ± 15%	-12 ± 15%	-12 ± 15%
Vastus Lateralis (VasLat)	1 ± 1%	1 ± 1%	1 ± 1%	1 ± 1%	1 ± 1%
Vastus Intermedius (VasInt)	9 ± 18%	9 ± 18%	9 ± 18%	9 ± 18%	9 ± 18%
Vastus Medialis (VasMed)	-9 ± 19%	-9 ± 19%	-9 ± 19%	-9 ± 19%	-9 ± 19%
Tibialis Anterior (TibAnt)	228 ± 16%	379 ± 76%	405 ± 85%	486 ± 205%	517 ± 215%
Gluteus Medius (GluteMed)	0 ± 0%	0 ± 0%	0 ± 0%	0 ± 0%	0 ± 0%
Gluteues Maximus (GluteMax)	2 ± 4%	2 ± 4%	2 ± 4%	2 ± 4%	2 ± 4%
Gluteus Minimus (GluteMin)	0 ± 0%	0 ± 0%	1 ± 0%	1 ± 1%	1 ± 2%
Biceps Femoris Long Head (BiFemLH)	1 ± 1%	1 ± 1%	1 ± 1%	1 ± 1%	1 ± 1%
Semimembranosus (Semimem)	3 ± 18%	4 ± 18%	5± 18%	5± 18%	5 ± 18%
Lateral Gastrocnemius (GasLat)	-24 ± 24%	-18 ± 44%	-12 ± 70%	-10 ± 71%	4 ± 77%
Medial Gastrocnemius (GasMed)	-14 ± 9%	-20 ± 20%	-22 ± 22%	-25 ± 26%	-23 ± 25%
Soleus	-13 ±6%	-23 ± 12%	-31 ± 17%	-34 ± 20%	-36 ± 19%

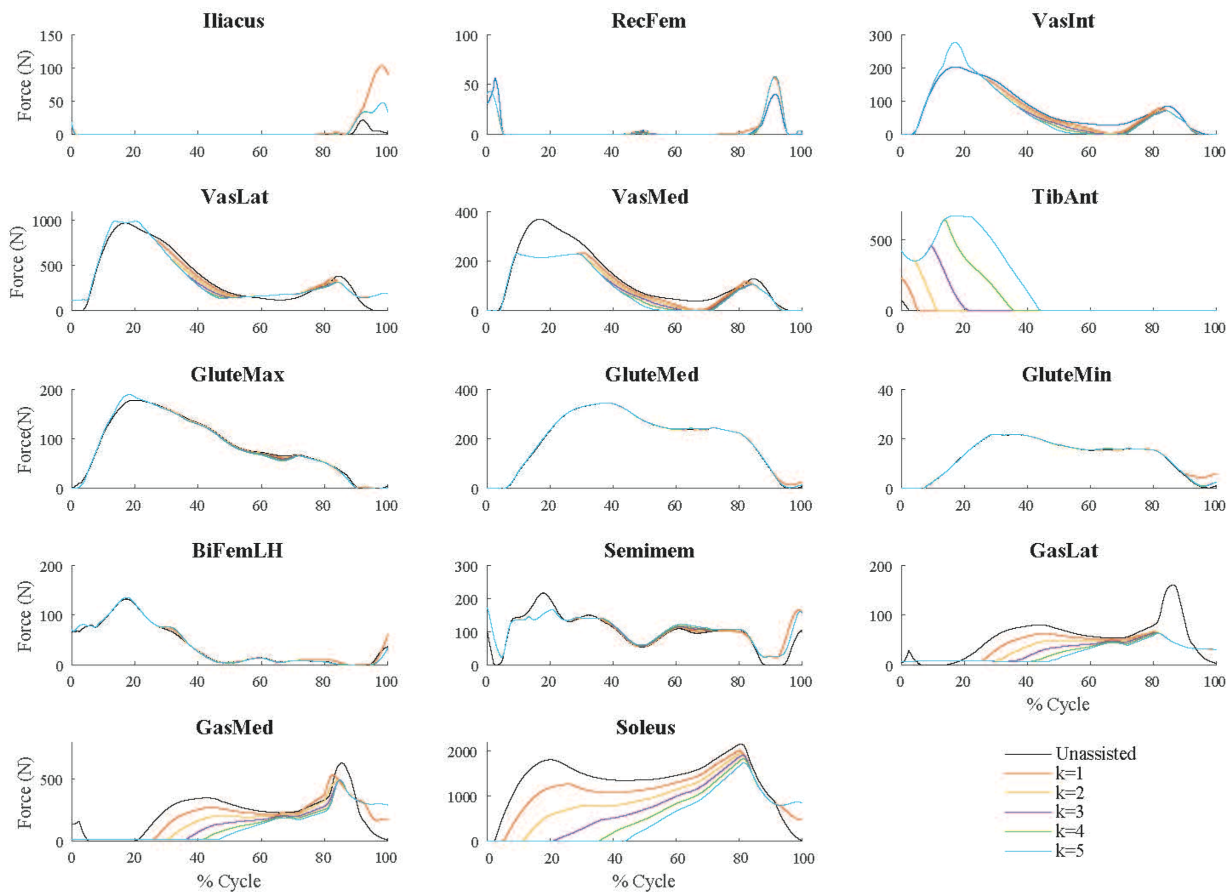

Figure 10: Selected muscle forces for one representative individual assisted at the ankle at varying spring stiffnesses in Nm/deg

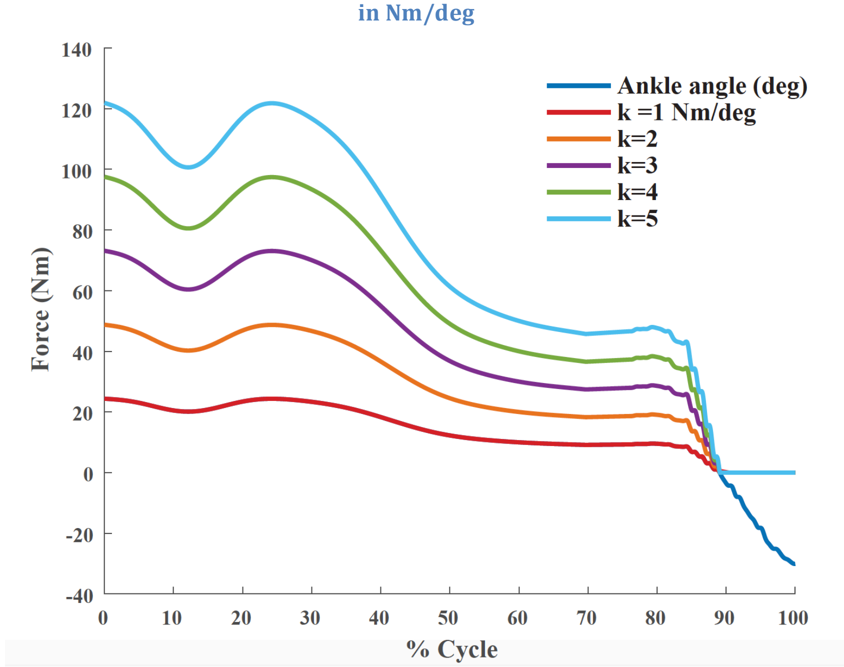

Figure 11: Experimental kinematics and spring forces at the ankle for one representative individual

When the knee was assisted, maximum force in the iliacus, rectus femoris, tibialis anterior, biceps femoris long head, semimbranosus, lateral gastrocnemius, and medial gastrocnemius increased (Figure10). However, maximum force in the vastus lateralis, vastus intermedius, vastus medialis, and soleus decreased. Maximum force in the iliacus increased more than other muscles in Table 7, while forces in the vastus intermedius decreased more than other muscles.

Table 7: Average % change in maximum force of selected muscles in individuals assisted at the knee

Stiffness (Nm/deg)	1	2	3	4	5
Iliacus (Iliac)	1306 ± 1076%	1564 ± 1071%	1614 ± 1160%	1641 ± 1141%	1641 ± 1141%
Rectus Femoris (RecFem)	552 ± 183%	310 ± 310%	128 ± 268%	92 ± 234%	-82 ± 24%
Vastus Lateralis (VasLat)	-77 ± 5%	-95 ± 10%	-93 ± 10%	-88 ± 15%	-92 ± 9%
Vastus Intermedius (VasInt)	-76 ± 4%	-100 ± 0%	-95 ± 10%	-85 ± 31%	-94 ± 7%
Vastus Medialis (VasMed)	-77 ± 5%	-100 ± 0%	-96 ± 8%	-88 ± 23%	-95 ± 6%
Tibialis Anterior (TibAnt)	378 ± 77%	390 ± 66%	415 ± 55%	431 ± 61%	431 ± 61%
Gluteus Medius (GluteMed)	14 ± 16%	53 ± 4%	115 ± 16%	129 ± 21%	125 ± 33%
Gluteues Maximus (GluteMax)	-14 ± 17%	-37 ± 22%	-33 ± 35%	-55 ± 52%	-61 ± 37%
Gluteus Minimus (GluteMin)	15 ± 12%	450 ± 297%	853 ± 70%	906 ± 29%	912 ± 34%
Biceps Femoris Long Head (BiFemLH)	506 ± 142%	424 ± 143%	590 ± 161%	748 ± 259%	753 ± 246%
Semimembranosus (Semimem)	614 ± 441%	633 ± 448%	633 ± 448%	634 ± 450%	642 ± 457%
Lateral Gastrocnemius (GasLat)	55 ± 132%	215 ± 240%	243 ± 260%	248 ± 268%	248 ± 268%
Medial Gastrocnemius (GasMed)	130 ± 99%	159 ± 101%	163 ± 108%	163 ± 108%	163 ± 108%
Soleus	-13 ± 9%	-32 ± 16%	-36 ± 19%	-36 ± 19%	-36 ± 19%

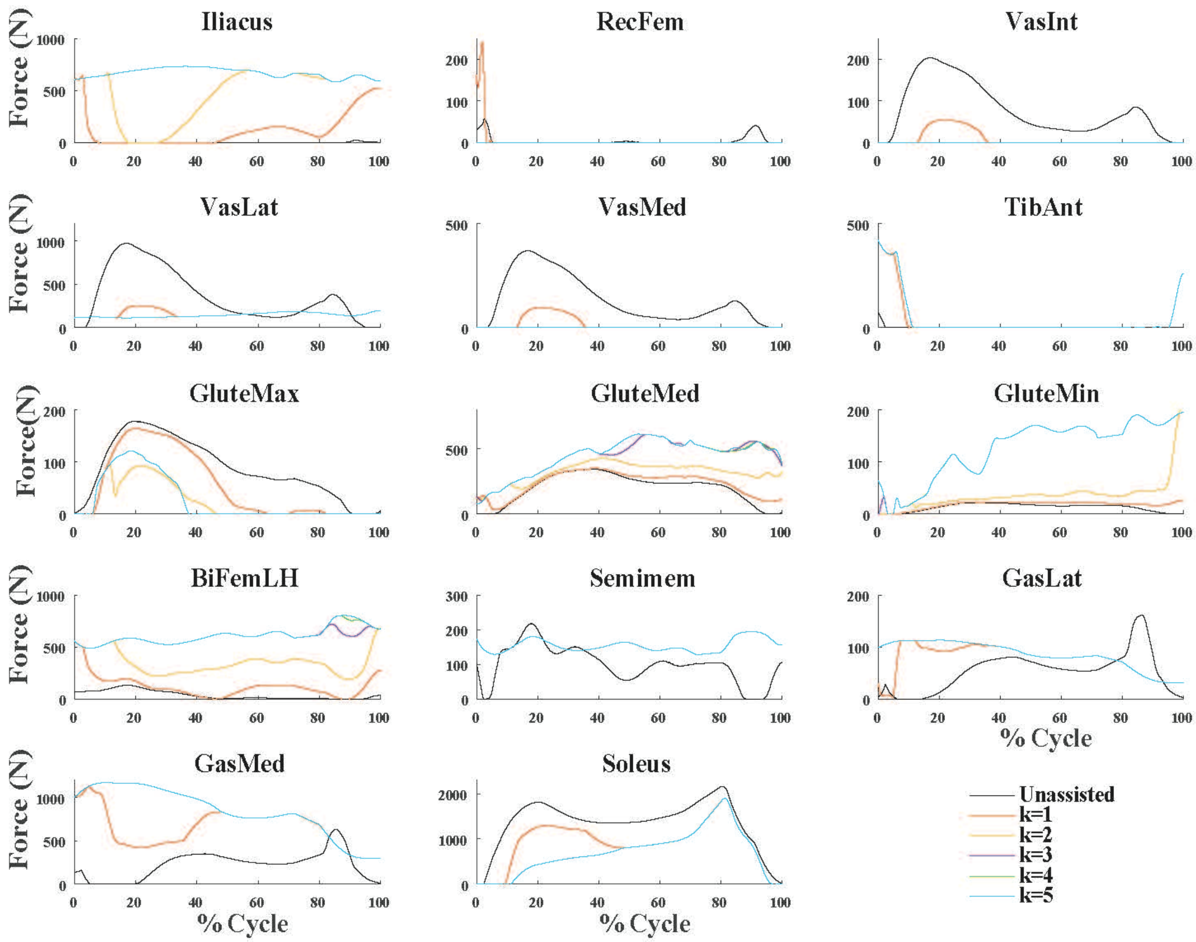

Figure 12: Selected muscle forces for one representative individual assisted at the knee at varying spring stiffnesses in Nm/deg

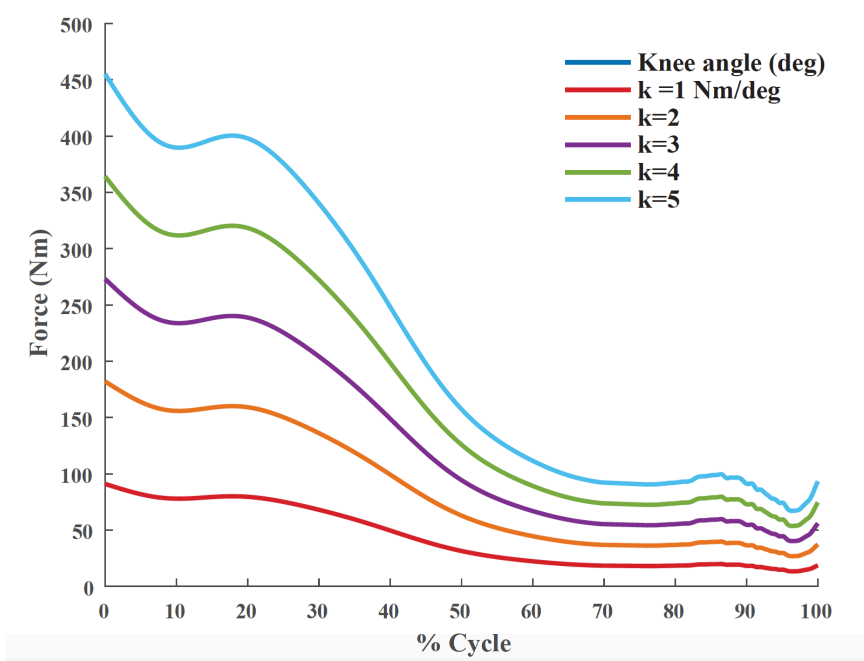

Figure 13: Experimental kinematics and spring forces at the knee for one representative individual. Knee angle and 1 Nm/deg spring forces are equal.

When the hip was assisted, maximum forces in the iliacus, rectus femoris, tibialis anterior, and medial gastrocnemius increased compared to the unassisted subject (Table 8). However, maximum forces in the vastus lateralis, vastus intermedius, vastus medialis, biceps femoris long head, semimembranosus, lateral gastrocnemius, and soleus decreased (Figure 11). Maximum force in the iliacus increased the most, while maximum force in the biceps femoris long head decreased the most. Forces in the biceps femoris long head decreased 100% for spring stiffnesses k = 2 and 3 Nm/deg.

Table 8: Average % change in maximum force of selected muscles in individuals assisted at the hip

Stiffness (Nm/deg)	1	2	3	4	5
Iliacus (Iliac)	715 ± 160%	1369 ± 1000%	1515 ± 985%	1622 ± 1127%	1641 ± 1141%
Rectus Femoris (RecFem)	553 ± 221%	925 ± 521%	925 ± 521%	925 ± 521%	925 ± 521%
Vastus Lateralis (VasLat)	-26 ± 5%	-50 ± 5%	-53 ± 9%	-51 ± 8%	-50 ± 6%
Vastus Intermedius (VasInt)	-25 ± 5%	-50 ± 5%	-53 ± 9%	-51 ± 8%	-33 ± 31%
Vastus Medialis (VasMed)	-30 ± 7%	-50 ± 5%	-54 ± 9%	-52 ± 8%	-40 ± 19%
Tibialis Anterior (TibAnt)	235 ± 45%	343 ± 111%	351 ± 100%	348 ± 110%	349 ± 103%
Gluteus Medius (GluteMed)	-9 ± 4%	-10 ± 5%	-10 ± 16%	-3 ± 27%	13 ± 28%
Gluteues Maximus (GluteMax)	-59 ± 12%	-90 ± 10%	-98 ± 3%	-96 ± 7%	-99 ± 1%
Gluteus Minimus (GluteMin)	-4 ± 5%	62 ± 93%	264 ± 196%	792 ± 118%	847 ± 78%
Biceps Femoris Long Head (BiFemLH)	-93 ± 11%	-100 ± 0%	-100 ± 0%	-90 ± 19%	-99 ± 2%
Semimembranosus (Semimem)	-25 ± 19%	-47 ± 30%	-65 ± 30%	-80 ± 27%	-90 ± 19%
Lateral Gastrocnemius (GasLat)	-13 ± 22%	-3 ± 24%	14 ± 42%	30 ± 64%	44 ± 88%
Medial Gastrocnemius (GasMed)	18 ± 38%	45 ± 58%	50 ± 56%	69 ± 44%	82 ± 49%
Soleus	-3 ±2%	-6 ± 5%	-9 ± 8%	-11 ± 9%	-11 ± 9%

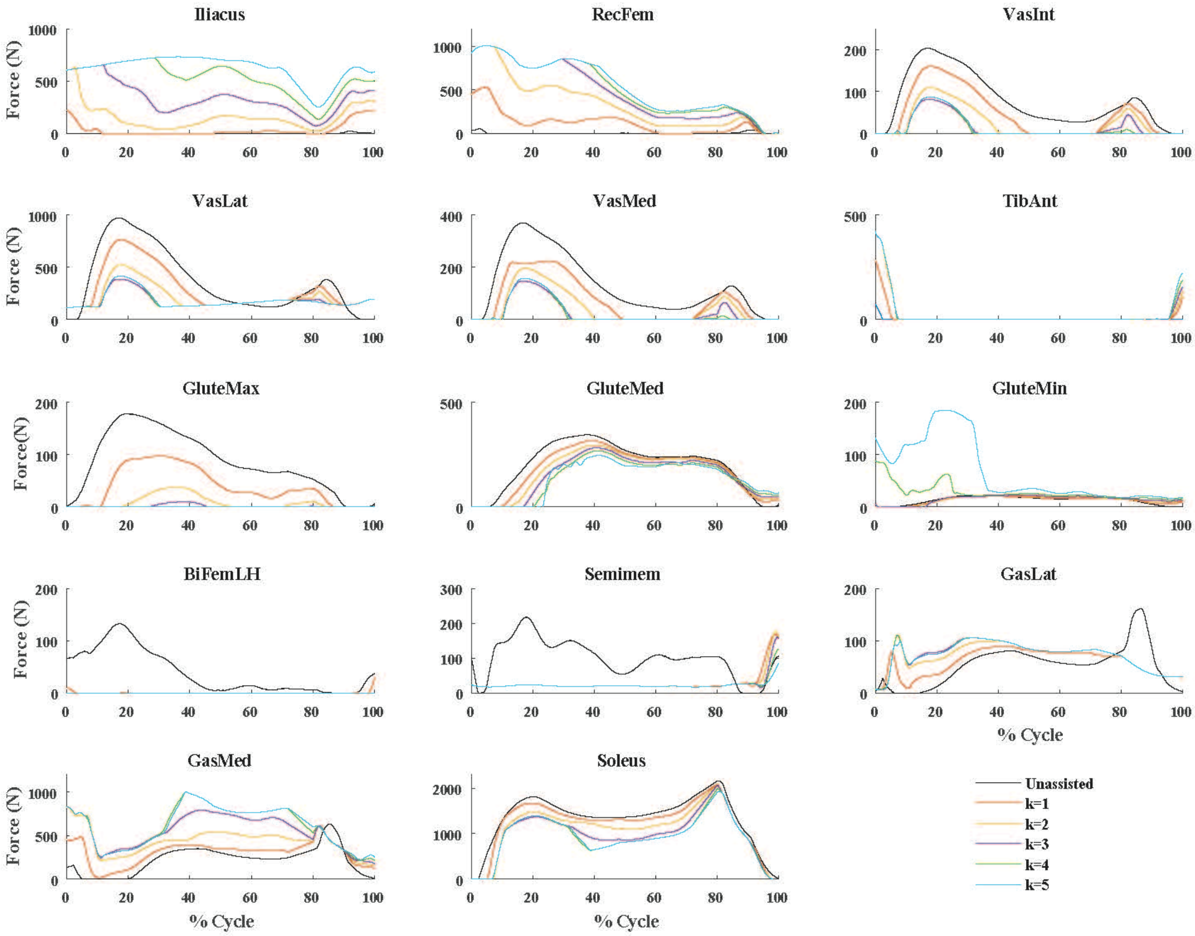

Figure 14: Selected muscle forces for one representative individual assisted at the hip at varying spring stiffnesses in Nm/deg

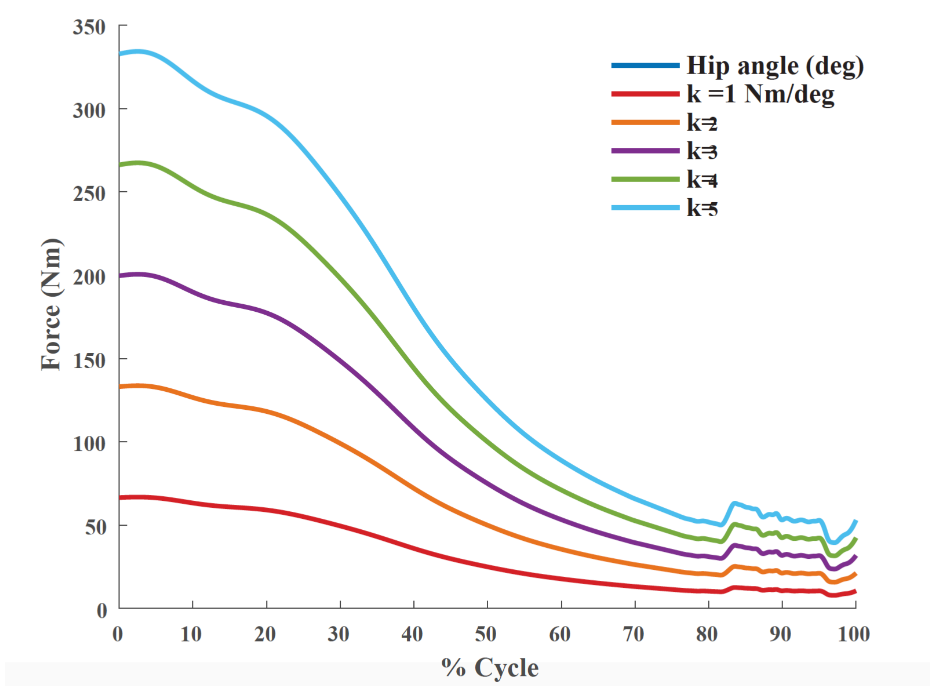

Figure 15: Experimental kinematics and spring forces at the hip for one representative individual. Hip angle and 1 Nm/deg spring forces are equal.

4. Discussion

4.1 General Conclusions

A spring with stiffness k = 1 Nm/deg located at the ankle had the lowest metabolic cost of the springs simulated. However, the unassisted individual had the lowest metabolic cost of all simulations. Overall metabolic cost increased for all spring stiffnesses at all joints. Even though overall metabolic cost increased for all cases, metabolic cost and forces decreased for certain muscles depending on which joint was assisted. Metabolic cost increased in the iliacus, tibialis anterior, medial gastrocnemius, and gluteus minimus for all cases. However, metabolic cost decreased in the vastus lateralis, vastus intermedius, vastus medialis, soleus, and gluteus maximus for all cases.

4.2 Assisted Stair Ascent Metabolic Cost Compared to Running and Gait

Metabolic cost increased for all variations of spring stiffness during stair ascent. However, previous studies of assisted running [19] and gait [21] showed metabolic cost decreased when individual were assisted. When an assistive device was located at the ankle during running at 2 m/s, the metabolic cost decreased 26% [19]. An unpowered, clutch-and-spring hip exoskeleton reduced the metabolic cost of gait by 10.3% [21]. Possible reasons for this include the use of passive devices instead of active devices like motors. Previous studies incorporated the assistive devices into the optimization objective function to predict the assistive device's torques [19]. While this may produce a less metabolically expensive assistive device, the resulting device may be costly to build due to the expense and time of designing and constructing such a device. The resulting device would be expensive and heavy, as it would likely consist of motors, batteries, and control systems. We used passive springs because they are inexpensive and can be bought off-the-shelf.

An increase in metabolic cost could also be caused by the limitation of tracking the experimental kinematics for the assisted simulations. An assistive device would most likely alter the kinematics of a person climbing stairs and potentially change the metabolic cost. Even though our results showed overall metabolic cost increasing contrary to previous studies, our study showed a decrease of metabolic cost in certain muscles.

4.3 Limitations

Experimental kinematics and kinetics were used as inputs to Static Optimization with the assistive device. Assistive devices would likely change the kinematics of an individual. However, the experimental kinematics provided a simpler way to predict metabolic cost changes of the assisted individuals instead of forward dynamics. It is possible that there would be a reduction in metabolic cost when the limitations of this study are removed. Using experimental kinematics of an individual with the assistive device instead of experimental kinematics of an unassisted individual could lead to a reduction of metabolic cost. Static Optimization was used which does not model the excitation-activation or tendon dynamics, which could also influence muscle forces. The torsional springs were also modeled as ideal and massless. A prototyped device would have mass associated with it that would influence the metabolic cost. Since the springs increased metabolic cost, the name assistive devices does not reflect the effect these simulated springs had on the individuals as they have to work harder when the device was added.

5. Conclusions

5.1 Contributions

To our knowledge, this is the first study to simulate passive assistive devices on older adults ascending stairs. The results from this study can better inform how to design assistive devices. In addition this study shows that it is less metabolically expensive to locate the assistive device at the ankle, while current devices focus on the hip like the Honda Walking Assist Device [10].

5.2 Future Work

Next steps include simulating the assistive devices on individuals with weakened muscles to better understand the impact of assistive devices on older adults. Simulations of older adults with simulated muscles weakness can better inform how assistive devices change compensation strategies for muscle weakness. Anne Marie Jackson found that when she simulated muscle weakness on older healthy women, stair ascent was most sensitive to ankle plantarflexor weakness [15]. When the ankle plantarflexors were weakened, the lateral and medial gastrocnemius and soleus showed increased activations [15]. Investigating the effects of the assistive devices from this study on the models with simulated muscle weakness can illustrate how assistive devices affect the activations of the muscles that compensate for ankle plantarflexors.

Assistive devices can also be simulated on movements such as descending stairs, gait, and sit-to-stand as individuals would likely wear the assistive device for several types of movement. After simulations on weakened subjects and additional movements, the optimal assistive device could be built and tested. Prototypes of the assistive devices could then be tested on older adults ascending stairs. Motion data could be collected and experimental and simulated kinematics could be compared to provide better understanding of the effects of assistive devices. Experimental metabolic cost could be gathered from VO2 testing to determine the benefits of the assistive device as well.

In addition only one cost model [27] was used to estimate metabolic cost. Other cost models such as the Umberger model [28] could be investigated in future work. Additional variations of the 1 to 5 Nm/deg torsional springs could also be simulated such as altering the neutral angle of the spring from zero degrees and adding additional elements like a clutch.

Summary

This thesis examined the effects of various spring stiffnesses on four individuals ascending stairs. We found that metabolic cost increased for all spring stiffnesses at all joints. However, we found that metabolic cost decreased for the vastus lateralis, vastus intermedius, vastus medialis, soleus, and gluteus maximus for all variations of the assistive device. Additionally, we found that the spring with stiffness k = 1 Nm/deg located at the ankle was the least metabolically expensive spring simulated, while the unassisted individual had the lowest metabolic cost of all simulations.

References

1. *Key Findings: Prevalence of Disability and Disability Type Among Adults, United States-2013*. Aug 20, 2015; Available from: https://www.cdc.gov/ncbddd/disabilityandhealth/features/key-findings-community-prevalence.html.
2. *Nearly 1 in 5 People Have a Disability in the U.S., Census Bureau Reports*. July 25, 2012; Available from: https://www.census.gov/newsroom/releases/archives/miscellaneous/cb12-134.html.
3. McKinnon, N. B., et al., *Neuromuscular contributions to the age-related reduction in muscle power: Mechanisms and potential role of high velocity power training*. Aging Research Reviews, 2017. **35**: p. 147-154.
4. *Important Facts about Falls.* Available from: https://www.cdc.gov/homeandrecreationalsafety/falls/adultfalls.html
5. Rejeski, W. J., Mihalko, S.L. *Physical Activity and Quality of Life in Older Adults*. Journals of Gerontology, 2001. **56A**(11): p. 23-25.
6. Reeves, N., et al. *Older adults employ alternative strategies to operate within their maximum capabilities when ascending stairs*. Journal of Electromyography and Kinesiology, 2009. **19**(2): p. e57-e68.
7. Larsen, A., Sørensen, H., Puggaard, L., and Aagaard, P. *Biomechanical determinants of maximal stair climbing capacity in healthy elderly women*. Scandinavian Journal of Medicine & Science in Sports, 2009. **19**(5): p. 678-686.
8. Mustafaoğlu, R., Unver, B. and Karatosun, V. *Evaluation of stair climbing in elderly people*. Journal of Back and Musculoskeletal Rehabilitation, 2015. **28**(3): p. 509-516.
9. S.M. Reid, et al. *Relationship between stair ambulation with and without a handrail and centre of pressure velocities during stair ascent and descent*. Gait & Posture, 2011. **34**: p. 529-532.
10. *Stair Lifts*. Available from: https://www.usmedicalsupplies.com/Stair-Lifts.htm.
11. *EZ-STEP*. Available from: http://www.ez-step.com/home.html.
12. *Honda to Begin Lease Sales of Honda Walking Assist Device*. July 21, 2015. Available from: http://world.honda.com/news/2015/p150721eng.html.
13. Sasaki, K., Eguchi, Y., and Suzuki, K. *A Wheelchair with Lever Propulsion Control for Climbing Up and Down Stairs*. Conf Proc IEEE Eng Med Biol Soc., 2016: p. 3358-3361.
14. Delp, S.L., et al., *OpenSim: open-source software to create and analyze dynamic Simulations of movement.* Ieee Transactions on Biomedical Engineering, 2007. **54**(11): p. 1940-1950.
15. Jackson, A., *Using Simulation to Investigate How Muscle Weakness Affects Stair Ascent*, 2017. M.S. thesis, The Ohio State Univ., Columbus, OH.
16. Caruthers, E.J., et al. *Muscle Forces and Their Contributions to Vertical and Horizontal Acceleration of the Center of Mass during Sit-to-Stand Transfer in a Young, Healthy Population.* Journal of Applied Biomechanics, 2016. **32**(5): p. 487-503.
17. Thompson, J.A., et al. *Gluteus maximus and soleus compensate for simulated quadriceps atrophy and activation failure during walking.* Journal of Biomechanics, 2013. **46**(13): p. 2165-2172.

18. Schloemer, S.A., et al. *Age-Related Differences in Gait Kinematics, Kinetics, and Muscle Function: A Principal Component Analysis.* Annals of Biomedical Engineering, 2017. **45**(3): p. 695-710.
19. Uchida, T.K., et al. *Simulating Ideal Assistive Devices to Reduce the Metabolic Cost of Running*. PLOS ONE, 2016. **11**(9): p. 1-19.
20. Ong, C.F., Hicks, J.L., and Delp, S.L. *Simulation-Based Design for Wearable Robotic Systems: An Optimization Framework for Enhancing a Standing Long Jump*. TBME, 2014: p. 1-10.
21. Saadatzi, M. and Celik, O. *A Passive Hip Exoskeleton for Reducing Metabolic Cost of Walking: A Simulation Study*. 41st Annual Meeting of the American Society of Biomechanics: 2017.
22. Andriacchi, T. P., Alexander, E. J., Toney, M. K., Dyrby, C., & Sum, J. *A Point Cluster Method for In Vivo Motion Analysis: Applied to a Study of Knee Kinematics*. Journal of Biomechanical Engineering, 1998. **120**(6): 743-749.
23. Camomilla, V., Cereatti, A., Vannozzi, G., & Cappozzo, A. *An optimized protocol for hip joint centre determination using the functional method*. Journal of Biomechanics, 2006. **39**:1096-1106.
24. Dorn, T. *C3D Extraction Toolbox*. simtk.org, 2008.
25. Hicks, J. *CMC best practices*. OpenSim User's Guide, 2015.
26. Schloemer, S. A., ThomSilder, A., Thelen, D. G., & Siston, R. A. *Age-Related Differences in Gait Kinematics, Kinetics, and Muscle Function: A Principal Component Analysis*. Annals of Biomedical Engineering, 2016.
27. Van der Krogt et al. *How robust is human gait to muscle weakness.* Gait Posture, 2012. **36**(1): 113-119.
28. Umberger, B. R., Gerritsen,K. G. M., Martin P. E. *A model of human muscle energy expenditure*. Comput. Methods Biomech. Biomed. Engin, 2003. **6**: 99–111.