THE INTERACTION OF PASSIVE AND ACTIVE ANKLE EXOSKELETONS WITH AGE-RELATED PHYSIOLOGICAL CHANGES TO IMPROVE METABOLIC COST

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INTERACTION OF ANKLE EXOSKELETON ASSISTANCE WITH AGE-RELATED CHANGES IN PHYSIOLOGY TO REDUCE METABOLIC COST OF WALKING

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TABLE OF CONTENTS

ACKNOWLEDGEMENTS	**iv**
LIST OF TABLES	**vii**
LIST OF FIGURES	**viii**
LIST OF SYMBOLS AND ABBREVIATIONS	**xii**
SUMMARY	**xiv**
ChAPTER 1. Introduction	**1**
CHAPter 2. In Vivo submaximal force-length behavior of the Soleus across age	**4**
2.1 Introduction	**4**
2.2 Methods	**6**
2.2.1 Participants	7
2.2.2 Protocol	7
2.2.3 Passive FL curves	9
2.2.4 Maximal Activation FL curves	10
2.2.5 Submaximal Activation FL curves	11
2.2.6 Tendon Stiffness	11
2.2.7 Walking	12
2.2.8 Statistics	13
2.3 Results	**13**
2.4 Discussion	**15**
CHAPTER 3. Neuromechanics and Energetics of motor and Spring-like ankle exoskeleton assistance in Older adults	**22**
3.1 Introduction	**22**
3.2 Methods	**24**
3.2.1 Participants	24
3.2.2 Exoskeletons	24
3.2.3 Walking Trials	27
3.2.4 Habituation Session	27
3.2.5 Gait Mechanics Session	27
3.2.6 Steady-State Metabolic Energy Consumption Session	28
3.2.7 Biomechanics Measurements	28
3.2.8 Electromyography (EMG) Measurements	29
3.2.9 Metabolic Cost Measurements	29
3.2.10 Statistics	30
3.3 Results	**31**
3.3.1 Metabolic Cost	32
3.3.2 Ankle Dynamics	33
3.3.3 Hip Dynamics	35
3.3.4 Electromyography (EMG)	38

3.3.5 Power 38
3.4 Discussion 39

CHAPTER 4. Linking Calf Muscle-Tendon Properties to User Performance in Active and Passive ankle Exoskeletons 43
4.1 Introduction 43
4.2 Methods 45
4.2.1 Participants 45
4.2.2 Exoskeleton 47
4.2.3 Protocol 47
4.2.4 Muscle-Tendon Characteristics 48
4.2.5 Habituation 49
4.2.6 Gait Mechanics 49
4.2.7 Steady-State Metabolic Energy Consumption 50
4.2.8 Biomechanics Measurements 50
4.2.9 Metabolic Cost Measurements 52
4.2.10 Statistics 52
4.3 Results 53
4.4 Discussion 53

CHAPTER 5. Conclusion 64

REFERENCES 67

LIST OF TABLES

Table 1 – Metabolic Power: Average net metabolic power for all participants and group averages for young and older adults. 42

LIST OF FIGURES

Figure 1 - Hypotheses and Protocol: a) A schematic showing less stiff tendons with age shift muscle fascicles to shorter, less optimal lengths and higher activations to reach the same force output. b) The protocol for creating 0, 33, 66, and 100% activation force-length curves for each participant. c) A sample image of the gastrocnemius insertion into the Achilles tendon. Tendon stiffness calculated as the slope of the linear fit of the FL curve from 50-100%. Functional tendon stiffness calculated as the slope of the tangent line at the maximal force measured during walking and a 2nd order polynomial fit of the FL curve. d) Participants laid supine on a dynamometer with their knee bent to isolate the soleus. The ankle angle was set at 20 PF, 10 PF, 0, 15 DF, and maxDF to change fascicle length. The inset is a sample image of the soleus and a representative fascicle. e) The participants received torque feedback for the 100% contractions and EMG feedback for the 33 and 66% contractions. 8

Figure 2 - Tendon Characterization: a) Individual Achilles tendon force-length curves for a) young (n=9) and b) older (n=6) adults. c) Tendon stiffness and d) functional tendon stiffness across age. Both tendon stiffness measures were statistically different with age (* $p < 0.05$). 16

Figure 3 - Group Normalized FL Curves and Properties: Age group averaged, activation dependent soleus force-length curves normalized to optimal length and maximal force for a) young and b) older adults. The dashed line corresponds to the maximal force produced during walking for the young adults. Average optimal length of the activation dependent force-length curves for c) young and d) older adults. e) Average maximum force achieved across age. f) Average passive muscle strain at 20% of maximum force by age. 18

Figure 4 - Individual Absolute FL Curves: Participant specific, activation dependent force-length curves for young (black) and older (purple) adults. 19

Figure 5 - Individual Normalized FL Curves: Participant specific, activation dependent force-length curves normalized to optimal length and maximal force for young (black) and older (purple) adults. 20

Figure 6 - a) The protocol consisted of 3 sessions: habituation, biomechanics, and metabolics. To habituate to the assistance each participant received 3 x 7 min bouts with the spring and motor assistance. The biomechanics session had 6 min of habituation for each assistance type (spring, motor, exo no assistance) with 2 minutes of collection 26

for every condition. The metabolics session had 6 min of habituation for each assistance type with 6 minutes of collection for every condition. b) The dephy ankle exoskeletons and the average assistance for the spring (pink), motor low (light blue), motor medium (medium blue), and motor high (dark blue). c) A participant walking in the exoskeletons with motion capture, EMG, and a metabolic mask (left). A participant walking showing how the plantarflexion assistance of the exoskeleton can indirectly assist hip extension in late stance and hip flexion in early stance on the contralateral limb.

Figure 7 - The change in net metabolic power from exoskeleton no assistance condition for young (left) and older (right) adults. The right axis is the percent change from the exoskeleton no assistance condition. The motors outperformed the spring for both age groups. The medium motor and the high motor had the greatest reduction for the young and older adults, respectively. 32

Figure 8 - Ankle and Exoskeleton Dynamics and Kinematics: The average a) biological contribution to the ankle moment, b) the exoskeleton torque, c) the biological contribution to the ankle power, d) the exoskeleton power, and e) the ankle angle for young (left) and older (right) adults. The bar graphs are the corresponding average change from the exoskeleton no assistance condition over early (10-40%) and late (40-70%) stance for young (left) and older (right) adults. 34

Figure 9 - Hip Dynamics and Kinematics: The average a) biological contribution to the hip moment, b) the biological contribution to the hip power, and c) the hip angle for young (left) and older (right) adults. The bar graphs are the corresponding average change from the exoskeleton no assistance condition over early (10-40%) and late (40-70%) stance for young (left) and older (right) adults. 35

Figure 10 - Ankle EMG: Average a) tibialis anterior (TA), b) soleus (SOL), and c) lateral gastrocnemius (GAS) activation across a stride for young (left) and older (right) adults. The bar graphs are the corresponding average change from the exoskeleton no assistance condition over early (10-40%) and late (40-70%) stance for young (left) and older (right) adults. 36

Figure 11 - Hip EMG: Average a) rectus femoris (RF), b) biceps femoris (BF), c) gluteus medius (GMd), and d) gluteus maximus (GMx) activation across a stride for young (left) and older (right) adults. The bar graphs are the corresponding average change from the exoskeleton no assistance condition over early (10-40%) and late (40-70%) stance for young (left) and older (right) adults. 37

Figure 12 - Metabolic and Mechanical Power: The relationship between the change in net metabolic power from exoskeleton no assistance with a) exoskeleton, b) ankle, and c) hip biological positive power for young (circle, solid) and older (triangle, dashed) adults. A reduction in metabolic power coincided with an increase in ankle positive power and a reduction in hip positive power for young adults. A reduction in metabolic power coincided with an increase in exoskeleton and ankle positive power and a reduction in hip positive power for older adults. 40

Figure 13 - a) Participant walking with ankle exoskeletons and portable metabolic system. b) The average exoskeleton torque provided over a stride. c) Soleus fascicle from B-mode ultrasound recorded over the medial gastrocnemius (left) and the gastrocnemius-Achilles tendon muscle tendon junction (MTJ) (right) for calculating stiffness. d) Hypothesized mechanism for reduced tendon stiffness (purple) and ankle exoskeleton assistance (pink) on theoretical force-length curves at different activations (30, 50, 70%). e) Mathematical hypothesis for how complaint tendons would modify operating ranges on force-length curves when walking. f) Mathematical hypothesis for how exoskeleton assistance would modify operating ranges on force-length curves when walking. g) Mathematical hypothesis for how people with complaint tendons would benefit more from exoskeleton assistance by modifying the operating ranges on force-length curves when walking. 46

Figure 14 - Average net metabolic power while walking in the exoskeleton without assistance (black), springlike assistance (pink), and motorlike assistance (blue). The data points represent participants with the circles being young adults and triangles being older adults. The motor was statistically different from both no torque and springlike assistance ($p < 0.05$). 54

Figure 15 - Ankle Inverse Kinematics and Dynamics: The stride average a) ankle angle, b) biological contribution to the ankle moment, c) exoskeleton torque for all exoskeleton conditions, d) biological contribution to the ankle power, and e) the exoskeleton power with early (0-40% stride) and late stance (40-70% stride) averages. 55

Figure 16 - Soleus Muscle Tendon Dynamics: The stride average soleus a) fascicle force, b) fascicle power, c) activation, d) fascicle length, and e) fascicle velocity with early (0-40% stride) and late stance (40-70% stride) averages. Both fascicle force and soleus activation were statistically different in late stance (* $p < 0.05$). 57

Figure 17 - Force-Length and -Velocity Curves: a) group average 0, 33, 66, and 100% activation force-length curves and the group average operating 58

range while wearing exoskeletons with no assistance, springlike assistance, and motorlike assistance. c) figure a zoomed in to see the workloops. Up triangles represent heel strike and down triangles represent toe off. b) Normalized force-velocity curve with group average operating range while wearing exoskeletons with no assistance, springlike assistance, and motorlike assistance. d) figure b zoomed in to see the workloops. Up triangles represent heel strike and down triangles represent toe off.

Figure 18 - Metabolic Power and Muscle-Tendon Properties: The relationship between the change in net metabolic power from the exoskeleton without assistance and a) Achilles tendon stiffness, b) passive muscle strain from zero to 20% of maximal force, c) maximal force, and d) optimal length at maximal activation for young (circle) and older (triangle) adults. The line is the linear regression fit to the data and the equation is above each corresponding plot. No correlation was found between the metabolic response to the exoskeleton and the muscle-tendon properties ($p > 0.05$). 60

Figure 19 - Individual Absolute Force Length Curves: Absolute participant specific, activation dependent force-length curves with workloops for walking in exoskeletons without assistance (black), springlike assistance (pink), and motorlike assistance (blue). 62

Figure 20 - Individual Normalized Force Length Curves: Normalized participant specific, activation dependent force-length curves with workloops for walking in exoskeletons without assistance (black), springlike assistance (pink), and motorlike assistance (blue). 63

LIST OF SYMBOLS AND ABBREVIATIONS

YA	Young Adult
OA	Older Adult
FL	Force-Length relationship
FV	Force-Velocity relationship
L_0	Optimal Length
L	Length
F_{max}	Maximum Force
F_{walk}	Maximum Force Required to Walk
F	Force
ε_{mus}	Passive Muscle Strain
k_t	Tendon Stiffness
K_{t_func}	Functional Tendon Stiffness
AT	Achilles Tendon
MTU	Muscle Tendon Unit
MTJ	Muscle Tendon Junction
MVC	Maximum Voluntary Contraction
PF	Plantarflexion
DF	Dorsiflexion
maxDF	Maximum Dorsiflexion
EMG	Electromyography
TA	Tibialis Anterior
SOL	Soleus
GAS	Gastrocnemius

RF	Rectus Femoris
BF	Biceps Femoris
GMd	Gluteus Medius
GMx	Gluteus Maximus
NoTorq	Exoskeleton without Assistance
Bio	Biological
Hz	Hertz
Δ	Change
ANOVA	Analysis of Variance

SUMMARY

Difficulties with mobility were the most reported disability for those age 65 and over. It is well known that older adults are slower and less economical while walking compared to young. This is thought to be brought on by reduced ankle push off power and a redistribution of positive power generation to more proximal joints (e.g., hip). We believe muscle and tendon level changes create a structural bottleneck leading to these functional changes. Ankle exoskeletons have been shown to increase ankle push off, modify muscle-tendon dynamics, and reduce metabolic cost in young adults for a near immediate improvement in walking performance. There is a *critical gap* in understanding whether beneficial exoskeleton assistance strategies for younger adults will also benefit older adults and if so, what the underlying mechanism is that enables exoskeletons to reduce metabolic cost across age.

My aims yielded a greater understanding of how people interact with ankle exoskeletons to modify metabolic cost. In Chapter 2, we characterize the muscle and tendon properties of young and older adults. This work is in prep to be submitted to the Journal of Biomechanics. In Chapter 3, we demonstrate how young and older adults respond to spring and motorlike exoskeletons. This work is in prep to be submitted to the Institute of Electrical and Electronics Engineers Transaction on Neural Systems and Rehabilitation Engineering. In Chapter 4, we demonstrate how spring and motorlike exoskeletons interact with the underlying calf muscle and tendon properties. These outcomes can improve the design and control of ankle exoskeletons to improve the cost of walking across age, leading to greater mobility and improved quality of life.

CHAPTER 1. INTRODUCTION

Aging has become such a large issue that the World Health Organization created a call to action in 2015 to bridge the gap between intrinsic capacity and functional ability of older adults [1]. This is a large issue because the aging population is growing and expected to grow over 70 million in the United States alone [2]. The expense of a growing older adult population is the average annual health care costs for Medicare have been steadily growing for the last ten years [3]. The costs are not only expensive for the government but also for the individuals and their families. A 2018 report found older adults spend as much money on food as they do on healthcare [3].

Difficulties in mobility are the leading disability reported in older adults [2]. Difficulties in mobility can include more energy required to walk, slower walking speeds, greater incidence of falls, and lead to a poor quality of life [4]–[7]. Functional changes stem from reduced ankle push off power and moment and a redistribution of power generation from the ankle to the hip [8], [9]. In a study providing feedback to reduce push-off force in young adults also saw the distal-to-proximal redistribution [10]. The issue metabolically with relying on the hip, is the muscles have long fascicles and stiffer tendons requiring more energy [11]. Conversely, the ankles have short fascicles and compliant tendons storing and returning energy for less costly walking [12].

The functional changes are often linked to structural changes in the muscle and tendon. Aging is associated with a loss of muscle mass, strength, power, and force generating capacity of the muscle and reduced tendon stiffness [13]–[15]. To improve the muscle-tendon deficits, the field looked to resistance training. The resistance training was able to improve muscle strength, power, mass, and tendon stiffness, the results were mixed and showed few functional improvements [16], [17]. To see results with resistance training requires a minimum of 4 weeks for muscle changes and 8 weeks for tendon changes, and even longer to see significant improvements to restore function [17]. Due to the lengthy time commitments, many studies found low compliance making it difficult to see this as a reliable solution [18]. Instead, we need a quicker and more reliable solution.

Exoskeletons are a quick solution for older adults to regain function. Researchers have been steadily making improvements in reducing metabolic cost over the last ten plus years [19]. At the moment, we can produce reductions of 50% with whole limb exoskeletons and 23% with ankle exoskeletons in the real world [20], [21]. There are many types of exoskeletons including passive, active, and pseudo-passive. Passive exoskeletons rely on springs and clutches to store and return energy while walking. Active devices rely on motors and batteries to provide torque. Pseudo-passive have small motors and batteries to modify springs for multiple stiffness options. Springlike assistance at the ankle typically provides a slow increase in torque through most of stance and peaks just before push off [22], [23]. Motorlike assistance at the ankle that has been optimized for metabolic cost by tuning assistance parameters provides a short and impulsive torque just before push off [24].

While exoskeletons have shown major improvements for young adults and other populations such as stroke, spinal cord injury, and cerebral palsy, there has been minimal testing with older adults [25]–[27]. Two groups tested hip exoskeletons but found small reductions of 3-7% [28], [29]. At the ankle a 4% reduction was found compared to shoes, however the positive power provided by the exoskeleton was low compared to assistance for young adults [30]. With this knowledge, we set out to compare exoskeleton assistance tuned for young adults to older adults and the interaction with the calf muscle and tendon.

In the following work, we first examined how calf muscle and tendon properties change with age to know the starting structure of the participants. We then compared springlike and motorlike assistance in young and older adults and their ability to reduce metabolic cost. This allowed us to determine if exoskeletons tuned for young adults were also beneficial for older adults. Lastly, we combined the structural properties with exoskeletons to see how exoskeleton assistance responded to muscle and tendon properties. This work uses novel techniques and populations to dive under the skin to understand how ankle exoskeletons modify metabolic cost.

CHAPTER 2. IN VIVO SUBMAXIMAL FORCE-LENGTH BEHAVIOR OF THE SOLEUS ACROSS AGE

2.1 Introduction

The ankle muscle tendon units are a primary driver of locomotion performance. Specifically, the ankle plantar flexors provide power generation during walking [31]. Force production of the ankle plantar flexors is a primary determinant of the metabolic cost of walking [32]. Force production of the plantar flexors is based on muscle mechanics, force-length (FL) and force-velocity (FV) properties. The FL and FV relationships are curves providing force potential at static lengths or velocities. The force potential mediated by these properties are also influenced by tendon properties. During walking, the long in series elasticity of the Achilles tendon allows the muscle and joint to decouple, providing economic force production and power amplification [33]. With age, it is thought that changes in muscle and tendon structural properties create a structural bottleneck affecting locomotion performance making walking less economical [12]. While numerous muscle properties have been quantified, it remains unknown how aging impacts the FL properties of the soleus. We can't confirm how structural properties of the muscle and tendon may lead to worse economy until they have been characterized.

The range over which a muscle operates on its FL relationship critically influences locomotion economy [32]. During walking, the soleus fascicles operate at a narrow range of lengths near the peak of their FL curve [34]. The tendon slows operating velocities and maintains lengths on the ascending limb for improved economy. One caveat to our current understanding of FL relationships is that typically, they are taken at maximal activations, which are not functionally relevant to walking and other daily tasks. Since most daily tasks are submaximal, when we map FL changes onto maximal curves we miss shifts with length, especially if optimal length changes with activation. Comparative literature and work in healthy young adults, show muscle has a leftward shift of optimal length with increasing force [35]. There is some debate whether this shift in optimal length is due to the protocol for creating FL curves. When creating curves as a percentage of force, it may be that the optimal length is longer purely because the muscle is generating less force and needs to shorten less. When creating curves as a percentage of activation, the shift in optimal length is no longer seen representing an optimal overlap in actin and myosin at the same length [36]. During more demanding tasks that require higher activations, the leftward shift would keep muscles operating at relatively longer lengths [37]. Thus, there is a need for characterizing submaximal FL curves for functionally relevant tasks.

With age there is a loss of muscle mass, strength, power, and force-generating capacity [13]. To relate these to the shape of the FL curve, the loss in strength reduces the maximal force of the curve (Fmax). When walking at matched speeds, older adults operate at shorter, less optimal lengths [38]. If the fascicles are adapting and overall getting shorter, the FL curve should become narrower. Due to the loss of muscle mass and smaller physiological cross-sectional area we would expect the FL curves to be shorter vertically [39]. Combining these changes, the normalized FL curves would maintain their shape with age, but absolute curves would shrink in both directions. Lastly, older muscles stiffen due to connective tissue and have reduced bulging [40]. The stiffening could restrict movement of the fascicles, lessening the shift in L_0 with activation.

Additionally, tendons can interact with the muscles to modify the force potential of the contraction. In aging, tendon stiffness is reduced by up to 39-43% [14], [15], [41]. When walking with less stiff tendons the tendon will stretch more, requiring the muscle to compensate by operating at shorter, less optimal lengths [38]. To reach the same force requirement to walk, the muscle must activate more at the shorter length affecting walking performance (Figure 1a).

The purpose of the study was to examine soleus activation-dependent FL relationships in aging. We assessed soleus torque on a dynamometer with EMG feedback to reach multiple activations. We varied ankle angles to modify fascicle lengths. We evaluated optimal length of the submaximal curves to corroborate the leftward shift as seen in literature. We hypothesized that both young and older adults will have a leftward shift in optimal length with increasing activation for the soleus muscle fascicle. A corollary hypothesis was that the older adult cohort will have reduced maximal force, and a smaller shift in optimal length with submaximal activation due to the changes seen with age. We also evaluated tendon stiffness because of its ability to affect force potential and the changes seen with age. We hypothesized that the older adult cohort would have less stiff Achilles tendon in agreement with literature. This work is the groundwork for understanding the soleus muscle at nonmaximal activations.

2.2 Methods

2.2.1 *Participants*

Eleven healthy young (2 female, 9 male; age: 21.11 ± 1.96 years; height: 176.72 ± 10.72 cm; mass: 74.61 ± 17.19 kg) and six healthy older (five female, one male; age: 70.83 ± 5.42 years; height: 166.23 ± 2.45 cm; mass: 60.92 ± 4.24 kg) adults participated. Participants had no history of leg musculoskeletal injuries, pain, or balance impairments, and could walk for at least 60 minutes within a 90-minute time frame. All participants gave informed written consent approved by the Georgia Institute of Technology central Institutional Review Board.

2.2.2 *Protocol*

Upon arrival, participants were fitted in a dynamometer (Biodex Medical Systems Inc., USA) laying supine with their right knee flexed at 120° (Figure 1d). Flexing the knee has been shown to limit torque contribution from the gastrocnemius and isolate the soleus [42]. The foot was tied down with a rachet strap to a custom-built pedal to prohibit the heel from lifting during contractions [43].

Motion capture markers were placed on the head of the tibia, medial malleolus, head of the 1st metatarsal, two on the dynamometer (one at the axis of rotation and one easily visible during collections), and on the pedal at the halfway point of the foot. The dynamometer markers were used to adjust for a difference in the axis of rotation of the

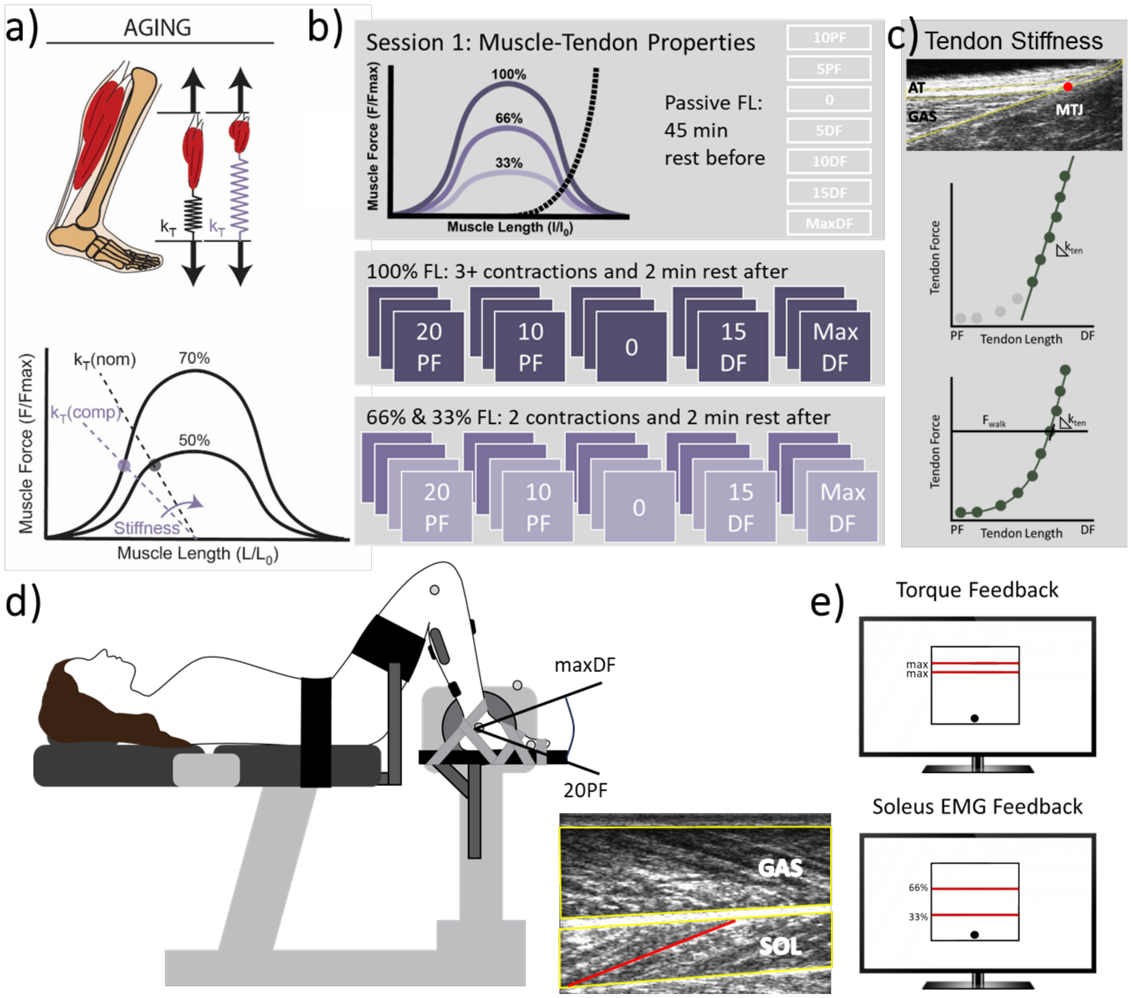

Figure 1 - Hypotheses and Protocol: a) A schematic showing less stiff tendons with age shift muscle fascicles to shorter, less optimal lengths and higher activations to reach the same force output. b) The protocol for creating 0, 33, 66, and 100% activation force-length curves for each participant. c) A sample image of the gastrocnemius insertion into the Achilles tendon. Tendon stiffness calculated as the slope of the linear fit of the FL curve from 50-100%. Functional tendon stiffness calculated as the slope of the tangent line at the maximal force measured during walking and a 2nd order polynomial fit of the FL curve. d) Participants laid supine on a dynamometer with their knee bent to isolate the soleus. The ankle angle was set at 20 PF, 10 PF, 0, 15 DF, and maxDF to change fascicle length. The inset is a sample image of the soleus and a representative fascicle. e) The participants received torque feedback for the 100% contractions and EMG feedback for the 33 and 66% contractions.

ankle about the dynamometer and to calculate moment arms. The moment arm from the ankle to the Achilles tendon was fixed and calculated by overlaying the ankle marker during a standing trial on to the ultrasound image from each participant and measured in ImageJ (Bethesda, MD) [44]. The marker on the pedal was placed at the halfway point of the foot to estimate the center of mass of the foot [45]. Surface electromyography (EMG) (Delsys Inc., Natick, MA) was placed on the medial soleus (Figure 1d). Soleus fascicle length was recorded with a linear-array B-mode ultrasound probe (Telemed, Vilnius, Lituani) secured over the medial gastrocnemius.

We collected motion capture data (200 Hz) (ViconMotion Systems, UK), torque on a dynamometer (1000 Hz) (Biodex Medical Systems Inc., USA), ultrasound (100 Hz), and EMG (1000 Hz) at the corresponding sampling frequencies. The EMG data was cleaned by removing the mean, bandpass filtered (20-450 Hz), notch filtered (60 Hz), absolute value rectified, and lowpass filtered at 6 Hz. Kinematics and kinetics were filtered at 6 Hz. Soleus fascicle lengths were measured using Ultratrack [46] (Figure 1d).

2.2.3 Passive FL curves

We captured still trials for 5 seconds to record passive muscle fascicle lengths via ultrasound and torque on the dynamometer. We performed a sweep of ankle angles from 10° plantarflexion (PF) to maximum dorsiflexion (DF) with 5° increments to change muscle fascicle length along the passive fascicle force-length curve (Figure 1b). Before each trial, a 45 second rest period occurred to mitigate elastic dissipative effects [34].

A custom MATLAB script (Mathworks Inc., Natick, MA) was used to calculate the soleus force. The gravitational torque from the weight of the pedal was removed from the measured torque. The resulting torque was then transformed from the axis of the dynamometer to the axis of the individual's medial malleoli. The torque from the weight of the foot was removed and then divided by the Achille's tendon moment arm to get the force of the muscle-tendon unit (MTU). The torque of the foot was estimated as 1.45% of body weight, multiplied by the moment arm between the estimated center of mass (halfway point of the foot) to the ankle [45]. The muscle-tendon unit force divided by the cosine of the pennation angle yielded passive muscle fascicle force [34]. An exponential equation was fit to the passive fascicle force-length curve [47]. Only trials with no EMG activity were processed. Muscle strain was calculated as the passive strain from zero at 20% of maximal force.

2.2.4 Maximal Activation FL curves

Maximum voluntary contractions (MVCs) were performed at 5 ankle angles (20PF, 10PF, 0, 15DF, maximum DF) in a random order, on the dynamometer with 2 minutes of rest between trials to minimize fatigue (Figure 1b, d). Participants were asked to slowly ramp up to their maximum effort and maintain that for 1-2 seconds before ramping down. On subsequent trials, the participants received visual feedback of their torque and were asked to beat their maximum from the previous trial (Figure 1e). Trials were repeated until they could no longer beat their maximum torque by more than 5% and had performed at least 3 trials at a given angle [48]. Trials were kept if participants reached at least 95% of the maximum activation, fascicle velocity was near zero, and ankle angular velocity was

<3 degrees per second. Active muscle fascicle force was calculated using the same process as outlined for the passive forces, except to capture only the force contributions of active muscle, the passive force at the contraction fascicle length was subtracted from the total force output.

Force-length curves were created using a custom MATLAB code, fitting a spline curve to 4 nodes. The 4 nodes were constrained to physiological values: L_0: minimum fascicle length found during maximal contractions as a lower bound and the maximum length during passive trials as an upper bound, Fmax: 1-1.1 times the active force during maximal contractions, onset of the ascending limb: 0.3-0.7 times L_0, and end of the descending limb: 1.3-1.8 times L_0. The curve fit selected reduced the root mean square error of the measured forces to the line.

2.2.5 Submaximal Activation FL curves

Visual feedback of soleus EMG activation was used to create submaximal force-length curves at 33% and 66% of the MVC at the same angles as the maximal curve (Figure 1b, e). Participants were asked to slowly ramp up to the desired activation and maintain that activation for 1-2 seconds before ramping down. Two contractions at each angle and activation level were performed with 2 minutes of rest between trials. Trials were checked for ±5% of the desired activation, velocity was near zero, and ankle angular velocity was <0.05 degrees per frame. The FL curve fits were calculated with the same methods as the maximal curves.

2.2.6 Tendon Stiffness

Achilles tendon stiffness measurements were calculated from plantarflexion contractions with a straight leg on the dynamometer (Biodex Medical Systems Inc., USA). The torque from the dynamometer was converted to muscle-tendon force using the Achilles tendon moment arm. We placed a B-mode ultrasound probe (Telemed, Vilnius, Lituani) over the muscle tendon junction (MTJ) of the gastrocnemius and the Achilles tendon (Figure 1c). Participants were asked to slowly contract as hard as possible. The data was parsed into 10% force intervals to track the length change in ImageJ (Bethesda, MD). A linear regression was fit to the force and length data in the 50-100% range of force output [49], [50]. The slope of the line is the stiffness in N/mm. The same data were used to calculate functional stiffness with a 2^{nd} order polynomial fit. A tangent line was created with the 2^{nd} order polynomial at the point of peak ankle force during walking and the slope is the functional stiffness.

2.2.7 Walking

Young participants walked at 1.25 m/s while instrumented with ultrasound placed over the medial gastrocnemius for two minutes. Kinematics were measured using a 10-camera motion capture system (100 Hz, Vicon, Oxford, UK) and a 44-marker lower limb and torso marker set. Ground reaction forces were captured with an instrumented split-belt treadmill (2000 Hz, CAREN, Motek, Netherlands). Inverse kinematics and inverse dynamics were calculated using OpenSim v4.0 (generic model 2392 with scapula) [51]. Joint angle data was filtered at 6 Hz (4^{th} order zero-phase butterworth) and forces were filtered at 15 Hz (4^{th} order zero-phase butterworth) prior to inverse dynamics calculations.

The soleus contribution to the ankle force was estimated as the ratio of the cross-sectional area of the soleus to the total cross-sectional area of the plantarflexor muscles (0.54) [52].

2.2.8 Statistics

Statistical analysis was performed in SPSS (IBM SPSS Statistics, Armonk, NY) with Laerd Statistics tutorials and guides. Tendon stiffness (Y: $n = 9$, O: $n = 5$) and functional tendon stiffness (Y: $n = 9$, O: $n = 5$) were measured to confirm the older cohort had less stiff tendons as seen in literature [14], [15], [41]. One older participant was removed due to the poor image quality. Independent samples t-test were performed with a Shapiro-Wilk test to test for normality of the data. Functional tendon stiffness for young adults did not pass the test of normality ($p > 0.05$), however t-tests are robust to deviations from normality. To test if there was a leftward shift in optimal length with increasing activation, a paired samples t-test and Shapiro-Wilk test of normality were performed. The difference in optimal length at 100 and 66% for older adults did not pass the test of normality ($p > 0.05$), however t-tests are robust to deviations from normality. To test for differences in maximal force, optimal length, and muscle strain with age, independent samples t-test was used with Shapiro-Wilk's test of normality. The older adult data for maximal force and the young adult data for muscle strain did not pass the test of normality ($p > 0.05$), however t-tests are robust to deviations from normality.

2.3 Results

All data reported will be mean ± standard deviation. To test the older adult cohort studied had reduced tendon stiffness as seen in literature, we measured tendon stiffness and functional tendon stiffness. A significant difference in tendon stiffness and functional tendon stiffness with age was found (Independent samples t-test: k_t: Y = 248.89 ± 75.88 N/mm, O = 162.20 ± 47.44 N/mm, p = 0.041; k_{t_func}: Y = 306.13 ± 154.12 N/mm, O = 152.58 ± 23.86 N/mm, p = 0.018) (Figure 2).

To test if both young and older adults have a leftward shift in optimal length with increasing activation for the soleus muscle individual force-length curves were created (Figure 4), normalized (Figure 5), and then averaged based on age group (Figure 3a, b). A significant difference in optimal length with decreasing activation for either age group was not found (Paired samples t-test: Y: $L_0^{100\%} - L_0^{66\%}$: p = 0.301, $L_0^{66\%} - L_0^{33\%}$: p = 0.630, $L_0^{100\%} - L_0^{33\%}$: p = 0.190; O: : $L_0^{100\%} - L_0^{66\%}$: p = 0.655, $L_0^{66\%} - L_0^{33\%}$: p = 0.563, $L_0^{100\%} - L_0^{33\%}$: p = 0.656) (Figure 3c, d).

To test if the older adult cohort have reduced maximal force and a smaller shift in optimal length with submaximal activation due to the changes seen with age optimal length for all activation curves and maximal force was measured. A significant difference in optimal length with age was not found(Independent samples t-test: $L_0^{100\%}$: Y = 38.78 ± 6.98 mm, O = 37.68 ± 10.83 mm, p = 0.814; $L_0^{66\%}$: Y = 39.90 ± 7.71 mm, O = 36.75 ± 11.11 mm, p = 0.526; $L_0^{33\%}$: Y = 41.00 ± 7.57 mm, O = 38.52 ± 10.60 mm, p = 0.604) (Figure 3a-d). A significant difference in maximal force with age was not found (Independent samples t-test: Y = 2430.54 ± 1312.86 N; O = 1630.10 ± 808.59 N p = 0.208) (Figure 3e). A significant difference in passive muscle strain at 20% maximal force with age was not found (Independent samples t-test: Y = 1.40 ± 0.30 L/L_0; O = 1.21 ± 0.13 L/L_0, $p = 0.183$) (Figure 3f).

2.4 Discussion

The purpose of the study was to examine in vivo soleus activation-dependent force-length relationships and tendon stiffness in aging. In agreement with previous studies, the older adults had lower tendon stiffness and functional tendon stiffness (Figure 2) [14], [15], [41]. There is mounting evidence in literature that the Achilles tendon becomes less stiff with age, but we wanted to ensure this was also seen in the cohort for this study. Functional tendon stiffness is the instantaneous stiffness at the average maximum force for young adults to walk. This shows that at matched force requirements for walking, the tendon is still less stiff for older adults.

The soleus activation-dependent force-length curves as a group look as expected (Figure 3). In both age groups, the forces go down with each submaximal activation as seen in other human and animal studies [36], [43], [53], [54]. Surprisingly, in contrast to our hypothesis, the maximal force of the older adults was not significantly lower from our young cohort (Figure 3e and 4). Literature has shown in multiple instances that maximal force is reduced with age [13]. The cohort of older adults used in this study were very active and healthy compared to older adults as a population. The high activity enabled us to define fascicles from ultrasound. Exercise is used to increase strength and by remaining active these older adults did not have reduced maximal forces.

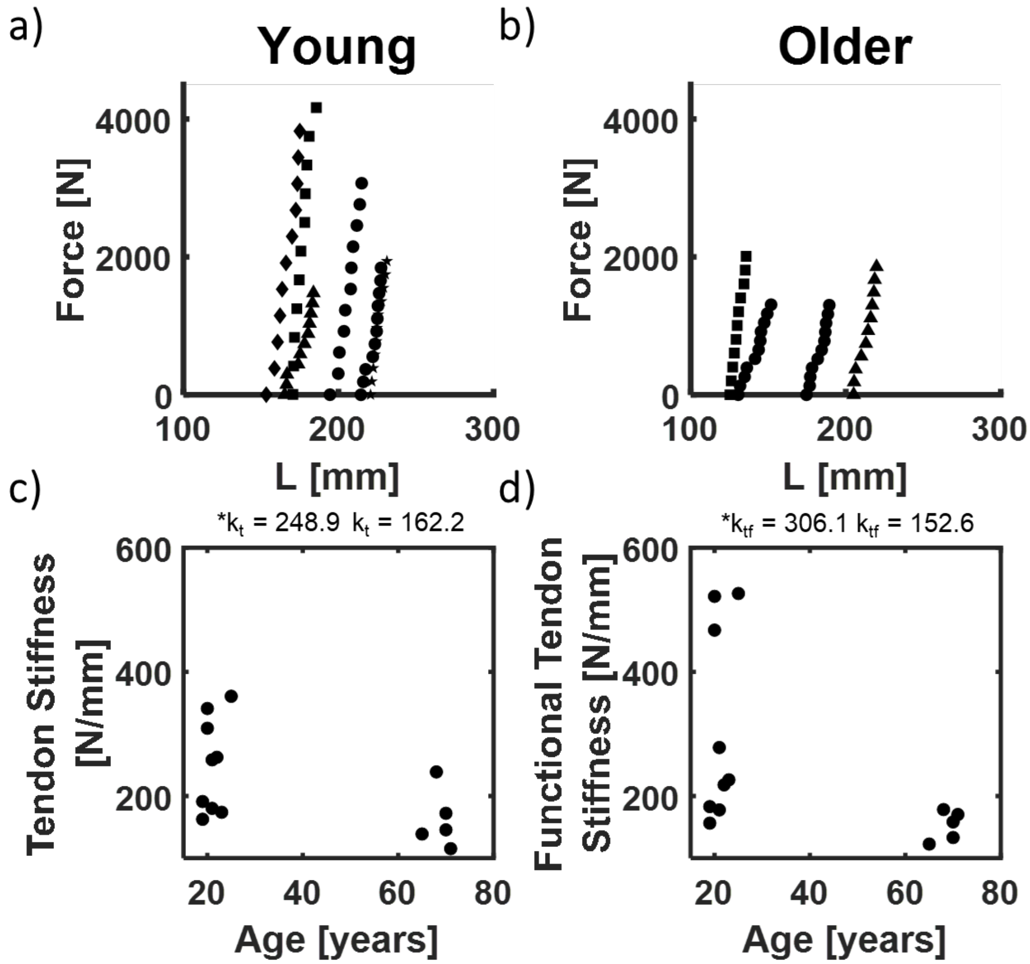

Figure 2 - Tendon Characterization: a) Individual Achilles tendon force-length curves for a) young (n=9) and b) older (n=6) adults. c) Tendon stiffness and d) functional tendon stiffness across age. Both tendon stiffness measures were statistically different with age (* $p < 0.05$).

It has been shown in animals and human vastus lateralis that there is a leftward shift in optimal length with decreasing force [36], [53], [54]. There is some debate whether this shift in optimal length is due to the protocol for creating the force-length curves. When creating curves as a percentage of force, it may be that the optimal length is longer purely because the muscle is generating less force and needs to shorten less. When creating curves as a percentage of activation, the shift in optimal length is no longer seen representing an optimal overlap in actin and myosin at the same length. Our data for the young cohort support the latter theory and are in opposition to our hypothesis that we would see a leftward shift with increasing activation. The optimal lengths across activation were not found to be different. However, in the plantarflexor muscles it is extremely difficult to get in vivo lengths on the descending limb. This has been shown in studies on the soleus, as well as the gastrocnemius [34], [43]. While our curve fits enforce a maximal force, a different curve may emerge if more data on the descending limb could be measured. To increase the likelihood of having data on the descending limb the participants went to their max dorsiflexion without pain or lifting the heel. I am unaware of an in vivo method that can provide more data for the descending limb.

Lastly, we hypothesized that the older adult cohort would have a smaller shift in optimal length with submaximal activation due to stiffening of the connective tissue in and around the muscle [40]. Since we didn't see meaningful changes in optimal length for either age group this hypothesis is no longer valid. We also tested muscle strain as a measure of muscle stiffness between the age groups, but they were not different. This measure better tests longitudinal stiffness, rather than axial stiffness so a measure that measures the 3D stiffness may show differences.

One limitation of the study is we chose active older adults to increase the likelihood of having better ultrasound to track fascicles. If we were able to perform this study in more sedentary older adults with lower maximal forces and compliant tendons there are functional implications for the metabolic cost of walking. For example, when matching the

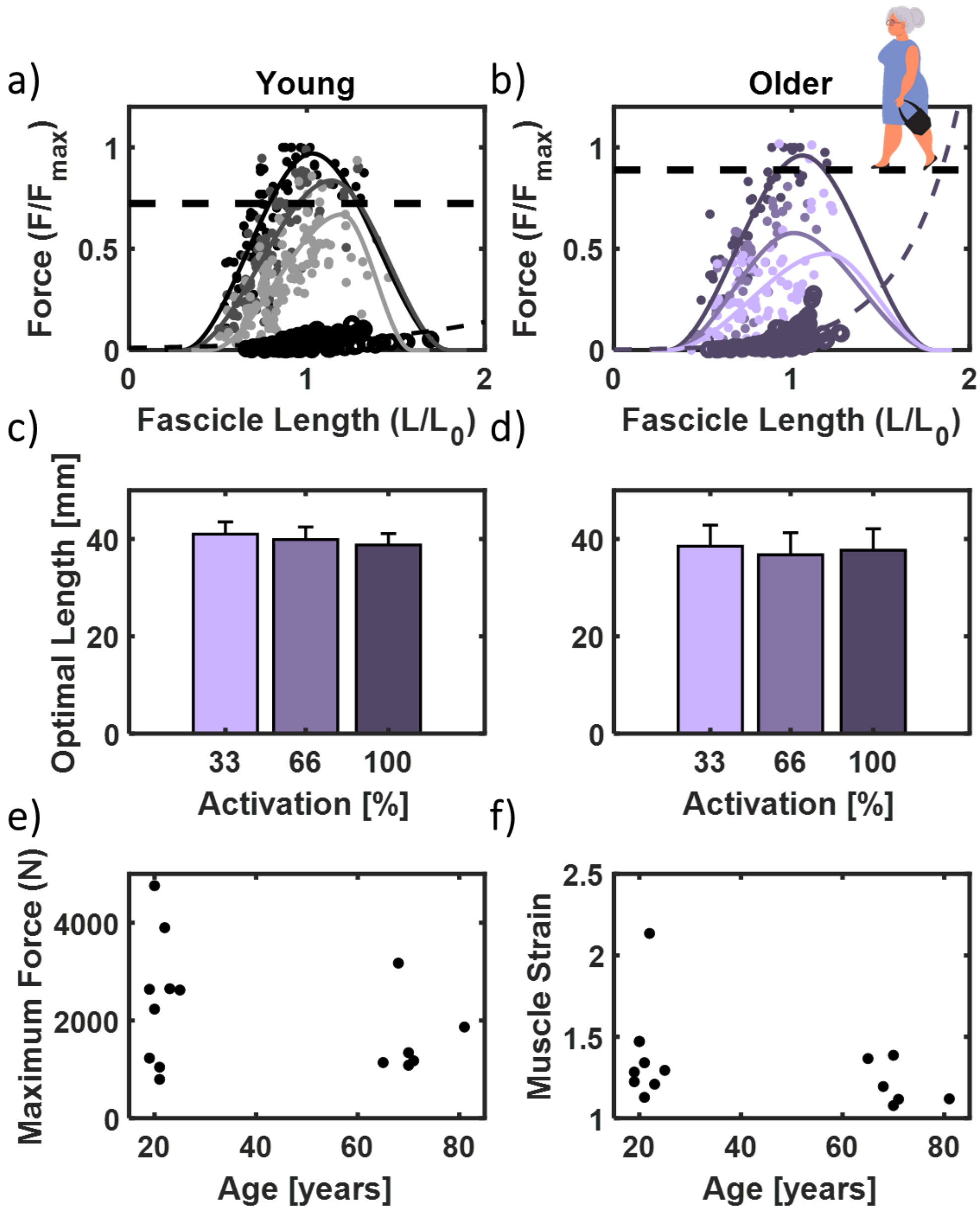

Figure 3 - Group Normalized FL Curves and Properties: Age group averaged, activation dependent soleus force-length curves normalized to optimal length and maximal force for a) young and b) older adults. The dashed line corresponds to the maximal force produced during walking for the young adults. Average optimal length of the activation dependent force-length curves for c) young and d) older adults. e) Average maximum force achieved across age. f) Average passive muscle strain at 20% of maximum force by age.

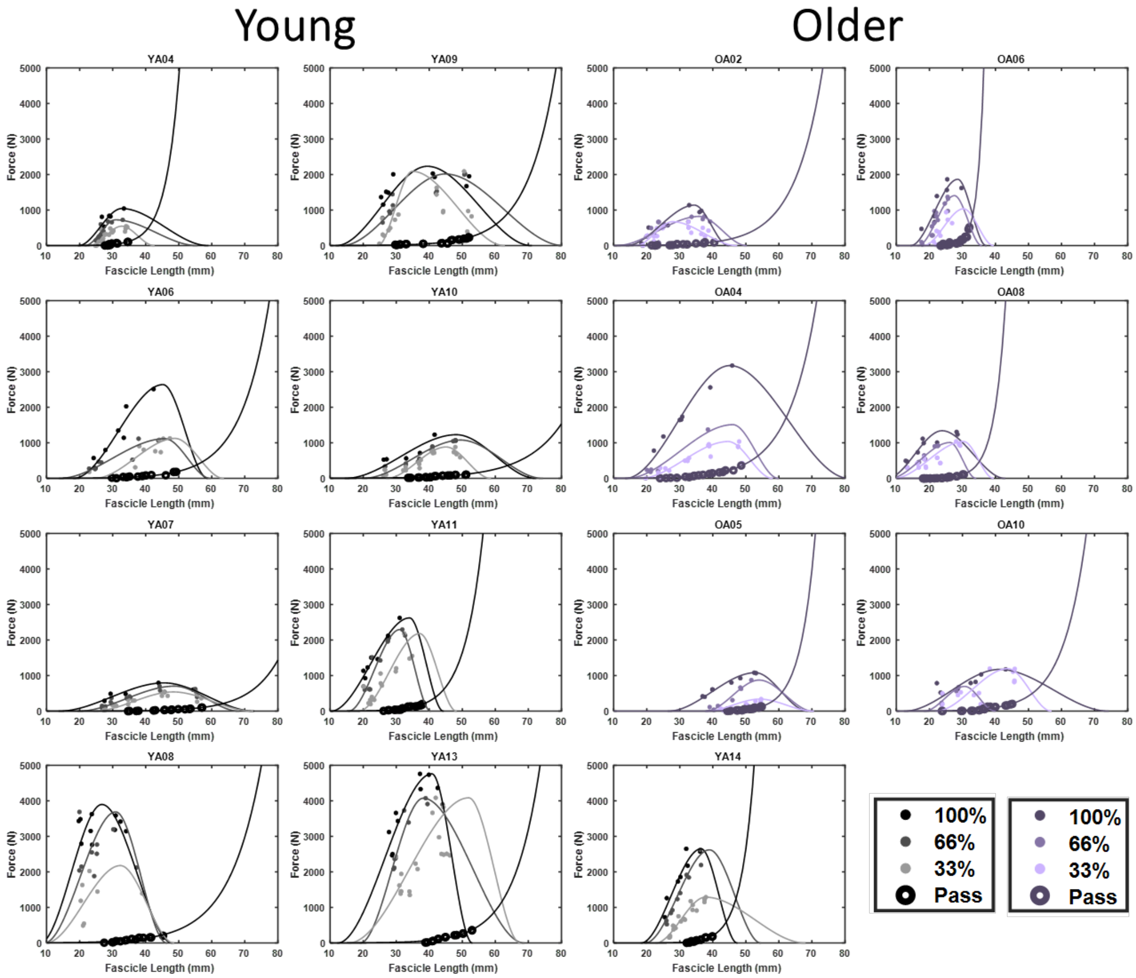

Figure 4 - Individual Absolute FL Curves: Participant specific, activation dependent force-length curves for young (black) and older (purple) adults.

maximal force required to walk (Figure 3a-b, dashed line), the older adults would be required to activate more to reach the same force leading to higher metabolic cost. The curves in Figure 3 suggest this may still be the case for the older adults in our study. Another limitation to our study is we did not electrically stimulate the muscle to ensure maximal contractions. It has been shown that it is hard to volitionally recruit muscle fibers at shorter lengths during a maximal contraction [55]. To reach the highest possible

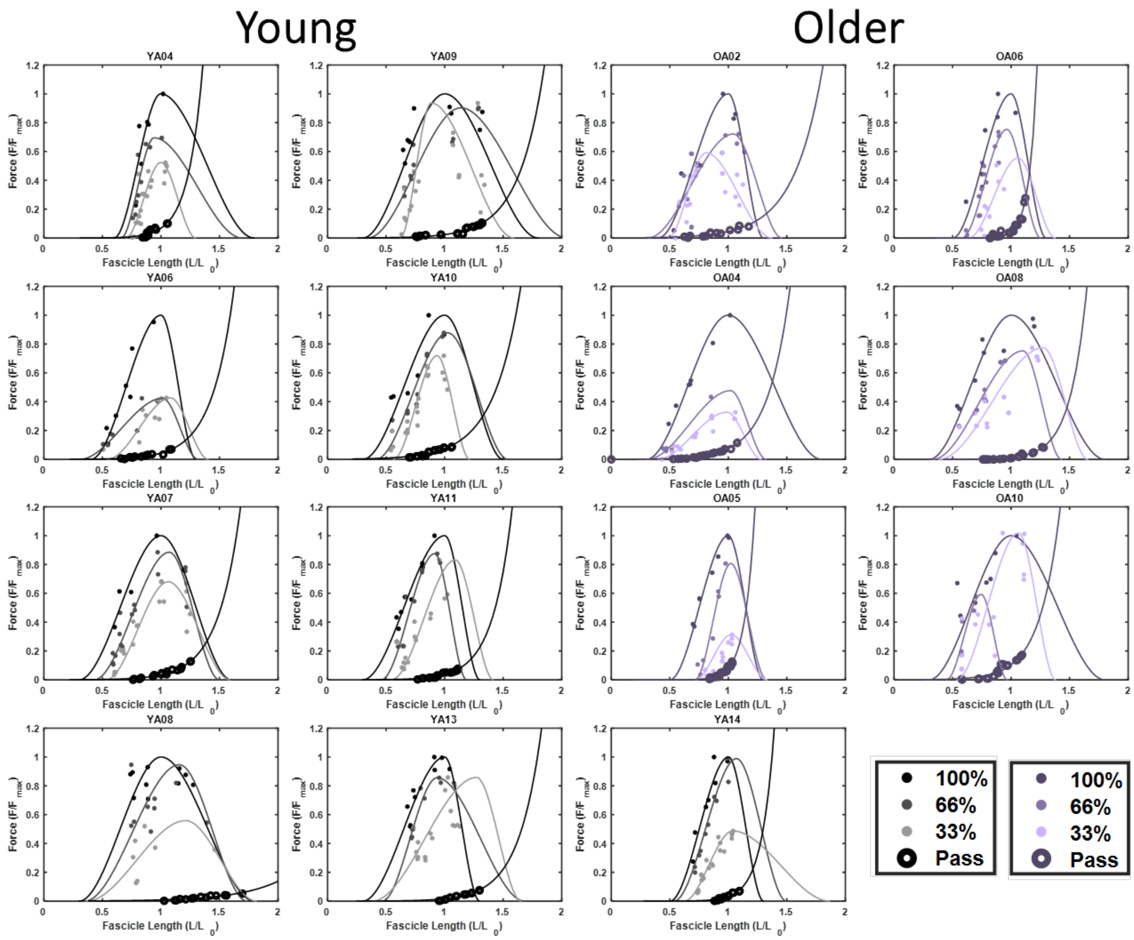

Figure 5 - Individual Normalized FL Curves: Participant specific, activation dependent force-length curves normalized to optimal length and maximal force for young (black) and older (purple) adults.

volitional contraction, we used torque feedback to encourage contracting as hard as possible.

In conclusion, while the older adults were more fit than the average population, we still saw reductions in tendon stiffness. Additionally, this is the first work to provide participant specific, activation dependent force length curves with age. This provides valuable data for linking physiological properties to functional performance. When combining reduced tendon stiffness with reduced maximal force typically found in older adults, there are functional implications that could explain higher metabolic cost of walking in older adults.

CHAPTER 3. NEUROMECHANICS AND ENERGETICS OF MOTOR AND SPRING-LIKE ANKLE EXOSKELETON ASSISTANCE IN OLDER ADULTS

3.1 Introduction

As we age, we walk slower and use more energy leading to a poor quality of life [4], [5]. A key driver of these changes is a reduction in mechanical power from the ankle [56], [57]. Mounting evidence in the literature indicates older adults have more compliant Achilles tendons, and this could disrupt an effective elastic mechanism that enables power amplification to generate impulsive peak power outputs at push-off [12]. This so-called structural bottleneck likely elicits a cascade of compensations that manifest a distal to proximal shift in mechanical energy generation [57]. This shift in mechanical effort comes with a metabolic penalty. Ankle joint mechanical work is done at high efficiency owing to significant elastic energy storage and return in the Achilles tendon [11]. Hip joint muscle-tendons have less compliant morphology and larger muscles that produce power less efficiently. To compensate for the new demand, older adults walk slower and less often, worsening their quality of life, making interventions highly sought after.

Interventions currently found in literature include eccentric exercise and strength training. While both of these have shown improvements in muscle mass, strength, power, and tendon stiffness, they have not consistently led to functional improvements for walking [16], [17], [58], [59]. Additionally, these interventions take at least 4 weeks for muscle and 8 weeks for tendon to see significant improvements and even longer to fully restore measures [60], [61]. On top of that, low compliance makes it an unreliable solution creating a critical need for acute and immediate solutions [18].

Exoskeletons are the assistive technology to provide immediate solutions for older adults [62]. In young adults, exoskeletons have gotten better at reducing metabolic cost over the past 15 years [19]. Exoskeletons about the lower limb joints have been shown to reduce metabolic cost by up to 50% in young adults [63]. Ankle exoskeletons alone, have been able to improve economy by 23% for young adults when walking out in the real world [21]. However, there has been limited testing of exoskeletons for older adults. There were two different torque profiles used for exoskeletons for older adults. The first behaves similarly to a spring applying torque slowly across most of stance. The second is more consistent with torque profiles applied by a motor with an impulsive peak in late stance. The spring-like profile saw reductions up to 19%, however the peak torque is much earlier than what has been optimized for young adults [64]. The motor-like profile saw 12% reductions from the added mass condition; however, the peak torque is lower than what has been optimized for young adults [30]. There is a *critical gap* in understanding whether ankle exoskeleton assistance strategies optimized for younger adults will also benefit older adults.

The purpose of this study was to explore ankle exoskeleton assistance strategies to reduce metabolic cost across the lifespan. One spring-like and three levels of motor-like assistance were tested for metabolic changes using indirect calorimetry. When optimizing spring and motor-like assistance for young adults, springs were able to reduce metabolic cost by 4.2-7.2% and motors by 23-34% [21]–[23], [63]. Based on this literature, we would hypothesize that motor-like exoskeleton assistance will be more beneficial than spring-like assistance regardless of age. With the distal-to-proximal shift seen in aging older adults use the less efficient hip joint. When adding ankle exoskeleton push off assistance, this can assist the ipsilateral ankle into plantarflexion and the ipsilateral hip into flexion in late stance, and the contralateral hip into extension in early stance (Figure 6c). This assistance, with the distal-to-proximal shift should allow for greater power at the ankle, offloading the less efficient hip. Therefore, we hypothesize ankle exoskeleton assistance would result in larger metabolic benefits for older versus younger adults regardless of assistance strategy (i.e., spring-like vs motor-like).

3.2 Methods

3.2.1 Participants

Ten healthy younger adults (3 female, 7 male; age: 21.70 ± 2.63 years; height: 175.58 ± 10.73 cm; mass: 72.57 ± 17.45 kg) and six healthy older adults (5 female, 1 male; age: 71.5 ± 5.54 years; height: 167.83 ± 6.69 cm; mass: 61.82 ± 9.39 kg) participated in the study. The participants met the inclusion criteria of not having a history of musculoskeletal injuries or vestibular impairments, and the ability to walk for a minimum of 60 minutes within a 90-minute time frame. All participants signed an informed consent approved by the Institutional Review Board at Georgia Institute of Technology.

3.2.2 Exoskeletons

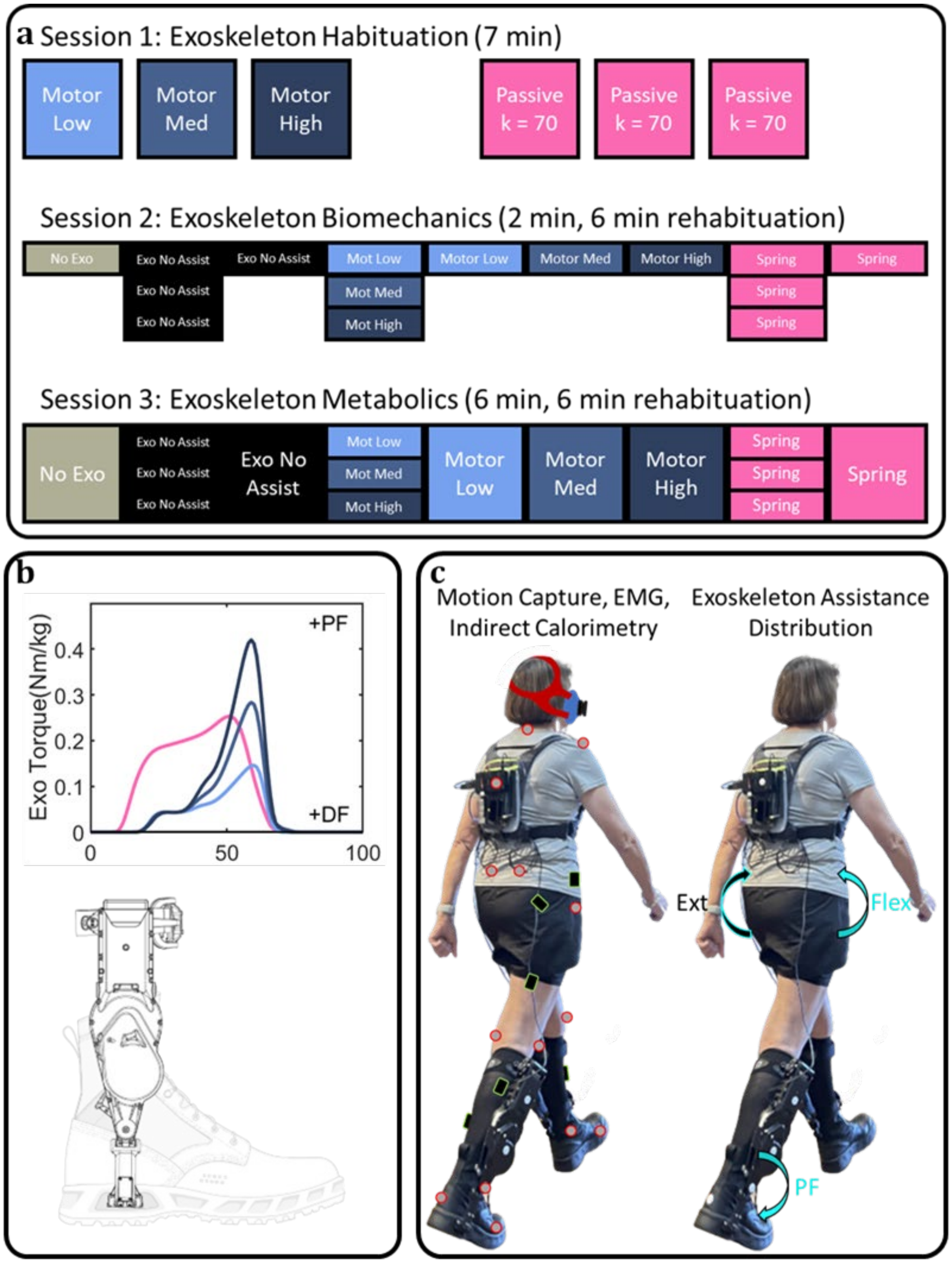
a
Session 1: Exoskeleton Habituation (7 min)
Motor Low
Motor Med
Motor High
Passive k = 70
Passive k = 70
Passive k = 70
Session 2: Exoskeleton Biomechanics (2 min, 6 min rehabituation)
No Exo
Exo No Assist
Exo No Assist
Exo No Assist
Exo No Assist
Mot Low
Mot Med
Mot High
Motor Low
Motor Med
Motor High
Spring
Spring
Spring
Spring
Session 3: Exoskeleton Metabolics (6 min, 6 min rehabituation)
No Exo
Exo No Assist
Exo No Assist
Exo No Assist
Exo No Assist
Mot Low
Mot Med
Mot High
Motor Low
Motor Med
Motor High
Spring
Spring
Spring
Spring
b
Exo Torque(Nm/kg)
0.4
0.3
0.2
0.1
0
0
50
100
+PF
+DF
c
Motion Capture, EMG, Indirect Calorimetry
Exoskeleton Assistance Distribution
Ext
Flex
PF

Figure 6 - a) The protocol consisted of 3 sessions: habituation, biomechanics, and metabolics. To habituate to the assistance each participant received 3 x 7 min bouts with the spring and motor assistance. The biomechanics session had 6 min of habituation for each assistance type (spring, motor, exo no assistance) with 2 minutes of collection for every condition. The metabolics session had 6 min of habituation for each assistance type with 6 minutes of collection for every condition. b) The dephy ankle exoskeletons and the average assistance for the spring (pink), motor low (light blue), motor medium (medium blue), and motor high (dark blue). c) A participant walking in the exoskeletons with motion capture, EMG, and a metabolic mask (left). A participant walking showing how the plantarflexion assistance of the exoskeleton can indirectly assist hip extension in late stance and hip flexion in early stance on the contralateral limb.

Commercially available bilateral ankle exoskeletons (Dephy Inc, Boxborough, MA) with plantarflexion assistance were used with custom controllers. The exoskeletons have on-board encoders and inertial measurement units for state-based control. Participants walked in six conditions: (1) without ankle exoskeletons, (2) exoskeletons without assistance, (3) exoskeletons with springlike assistance, and exoskeletons with three motorized assistance strategies with increasing peak torque ((4) 10, (5) 20, (6) 30 Nm). The springlike assistance used impedance control without damping to apply torque with a spring stiffness of 70 Nm/rad [23]. The assistance began at flat foot after heel strike and ended at toe off mimicking a spring and clutch mechanism to time when to release the stored energy. The torque profile for the motor assistance was a human-in-the-loop optimized spline generated from four nodes: peak time, rise time, fall time, and peak torque [24] (Figure 6b). The parameters used were optimized for metabolic cost with the average younger adults' parameters from Zhang et al. The only difference was peak torque was preset to 10, 20, and 30 Nm for this experiment.

3.2.3 *Walking Trials*

Participants completed testing over three sessions walking at 1.25 m/s on an instrumented split-belt treadmill (CAREN, Motek, Netherlands). The sessions were conducted in the order of 1) habituation, 2) gait mechanics, and 3) steady-state metabolic energy consumption. Waiting periods between sessions were minimized to allow for learning and retention (2-7 days on average).

3.2.4 *Habituation Session*

Previous work has shown the importance of habituation for the acceptance of assistance from exoskeletons [65]. In the first session, each participant walked with spring-like assistance for three, seven-minute intervals for a total of 21 minutes. They also walked with motorized assistance for three, seven-minute intervals while increasing peak torque (10, 20, 30 Nm) with each bout.

3.2.5 *Gait Mechanics Session*

The primary goal for the second session was to obtain biomechanical measurements. We instrumented participants with surface electromyography to record muscle activity on the right leg [tibialis anterior (TA), lateral gastrocnemius (LG), soleus (SOL), rectus femoris (RF), biceps femoris (BF), gluteus medius (GMd), and gluteus maximus (GMx)] and reflective markers on the lower limbs and torso to collect motion capture. Participants were rehabituated to the exoskeleton assistance for 6-minutes prior to walking with exoskeletons without assistance, with springlike assistance, and with motorized assistance. During the motorized assistance habituation period, we increased the peak torque every 2 minutes. After each condition's habituation trial, participants walked for 2 minutes. The exoskeleton assistance order was randomized for each participant.

3.2.6 Steady-State Metabolic Energy Consumption Session

Participants began the third session with a five-minute standing trial to record their metabolic resting baseline. Participants were fitted to a portable indirect calorimetry device (COSMED K5, Rome, Italy) to collect energy expenditure for each condition. We repeated the habituation process conducted on the gait mechanics session. The participants walked for six minutes at each of the six conditions in a randomized order with the respective habituation trial before.

3.2.7 Biomechanics Measurements

Kinematics were measured using a 10-camera motion capture system (100 Hz, Vicon, Oxford, UK) and a 44 marker lower limb and torso marker set. Ground reaction forces were captured with an instrumented split-belt treadmill (2000 Hz, CAREN, Motek, Netherlands). Inverse kinematics and inverse dynamics were calculated using OpenSim v4.0 (generic model 2392 with scapula) [51]. Joint angle data was filtered at 6 Hz (4^{th} order zero-phase Butterworth) and ground reaction forces were filtered at 15 Hz (4^{th} order zero-phase Butterworth) prior to inverse dynamics calculations. The biological contribution to the total ankle joint moment was calculated by subtracting the estimated exoskeleton torque from the total ankle moment. Joint angles, moments, and powers were calculated for the sagittal plane at the ankle and hip. All data was separated by stride using a 30 N threshold of the vertical ground reaction force. The stride average is the average of five strides. Joint moments and powers were averaged over early (0-40 % stride) and late stance (40-70% stride) for each individual and then group averaged.

3.2.8 Electromyography (EMG) Measurements

Muscle activity of the tibialis anterior (TA), soleus (SOL), lateral gastrocnemius (LG), rectus femoris (RF), biceps femoris (BF), gluteus medius (GMd), and gluteus maximus (GMx) were measured using surface EMG on the right leg (2000 Hz, Avantis, Delsys, Natick, MA). The EMG locations on the skin were shaved and exfoliated for improved signal. The data was processed by removing the mean, 20-400 Hz bandpass filtered (8^{th} order zero-phase Butterworth), rectified using the absolute value, and smoothed using a 20 Hz lowpass filter (3^{rd} order zero-phase Butterworth). The filtered data was stride averaged and resampled to 2000 datapoints. The EMG amplitude was normalized to the peak for the given muscle from the filtered, stride-averaged EMG from the exoskeleton condition without assistance for each participant. The integrated EMG was computed as the time-integral of the EMG averaged across each stride divided by the time period over which it was integrated for early stance (0-40%) and late stance (40-70%).

3.2.9 Metabolic Cost Measurements

Indirect calorimetry was collected using a portable system (COSMED K5, Rome, Italy) as breath-by-breath measurements for each walking trial. Each trial lasted six minutes to ensure the steady-state metabolic rate was achieved. Metabolic power was calculated using Brockway's equation and averaging the last minute of the trial [66]. Net metabolic power was calculated by subtracting the standing trial average from the walking trial averages. The exoskeleton no assistance trial was then subtracted off to get the change and percent change in net metabolic power.

3.2.10 Statistics

Statistical analysis was only performed on the change from exoskeleton no torque condition (NoTorq) to isolate the influence of the exoskeleton assistance from the effect of the added mass. For change in net metabolic power, we reported the means and standard error calculated across participants. Change in net metabolic power was the primary outcome measure in this study. Based on previous literature, we expected the motorized assistance to outperform the springlike assistance. Therefore, we performed a two-way mixed ANOVA (IBM SPSS Statistics, Armonk, NY). There were no outliers, as assessed by examination of studentized residuals for values greater than ±3. Metabolic power data was normally distributed, as assessed by the Shapiro-Wilk's test ($p > 0.05$). There was homogeneity of variances ($p > .05$). and covariances ($p > .001$, $p = .687$)., as assessed by Levene's test of homogeneity of variances and Box's M test respectively. Mauchly's test of sphericity indicated that the assumption of sphericity was met for the two-way interaction, $\chi^2(2) = 14.59$, $p = .106$. To test the differences between exoskeleton conditions and exoskeleton conditions in each age group, a post-hoc pairwise comparison with Bonferroni correction was used. Statistics were performed based on Laerd Statistics tutorials and guides.

There was no statistically significant effect of age (p = 0.668). There was a statistically significant interaction between the exoskeleton condition and age group on the change in metabolic power ($F(4, 56) = 3.246$, $p = 0.018$ ($p < .05$, partial $\eta^2 = .188$). There was a statistically significant effect of exoskeleton condition on metabolic power for both the young and older groups, (Young: $F(4, 36) = 7.64$, $p < .001$, partial $\eta^2 = .459$, Older: $F(4, 20) = 17.22$, $p < .001$, partial $\eta^2 = .775$. For the young age group, metabolic power was statistically significantly reduced for the medium motor compared to the spring and the low motor assistance (Med: -0.731 ± 0.097 W/kg, Spring: -0.136 ± 0.122 W/kg p=0.005, Low: -0.352 ± 0.159 p = 0.022). For the older age group, metabolic power was statistically significantly increased for the spring compared to no exoskeleton, medium motor, and high motor (NE<spring, spring>S20 & S30). Metabolic power for each exoskeleton condition was not statistically significantly different between young and older.

3.3 Results

The springlike assistance condition (Spring) details what occurs with a passive exoskeleton that has shown small but significant reductions in metabolic cost for young adults. The high motor condition (Motor High) details what occurs with an active exoskeleton that was optimized for metabolic cost of young adults.

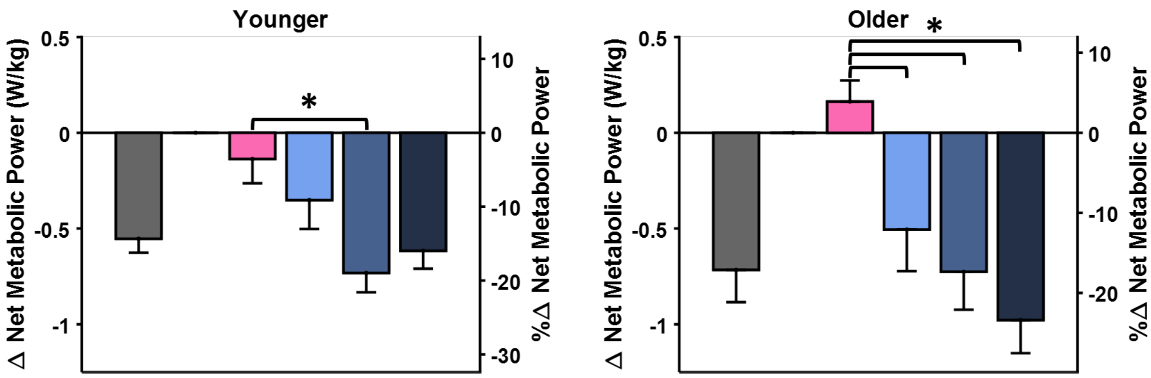

Figure 7 - The change in net metabolic power from exoskeleton no assistance condition for young (left) and older (right) adults. The right axis is the percent change from the exoskeleton no assistance condition. The motors outperformed the spring for both age groups. The medium motor and the high motor had the greatest reduction for the young and older adults, respectively.

3.3.1 Metabolic Cost

To test if motorized assistance will provide a greater metabolic benefit than springlike assistance, I measured net metabolic power for all conditions regardless of age groups (Table 1). In agreement with our hypothesis, the springlike assistance was statistically different from all motor conditions and walking without an exoskeleton. The exoskeleton no assistance condition was statistically different from the two higher motors and both motors were different from each other. The no assistance condition was different from the no exoskeleton condition as well. In conclusion, the two highest motors reduced metabolic cost, while the spring did not, from the no assistance condition. To test if older adults would receive a greater metabolic benefit regardless of assistance strategy, I measured the change and percent change in net metabolic power for all conditions and both age groups (Figure 7). In disagreement with our 2nd hypothesis there was no significant difference found between young and older adults across exoskeleton assistance. The springlike assistance had opposite effects for young and older adults, but neither were statistically different from no assistance. The greatest reduction in metabolic cost came with the medium motor for young adults and the high motor for older adults.

To further inspect the differences from exoskeleton no assistance, the change and percent change were calculated. The older adults had a larger reduction than young in net metabolic power with motor high compared to exoskeleton no assistance (Change Y: -0.616 ± 0.322 W/kg, O: -0.978 ± 0.489 W/kg; % Change Y: -15.967%, O: -23.384%).

3.3.2 Ankle Dynamics

To determine where these reductions in metabolic cost were coming from inverse kinematics and dynamics were calculated for the ankle and the hip (Figure 8, Figure 9 respectively). At the ankle both age groups reduced ankle moment at push off with all exoskeleton assistances (Figure 8a). The exoskeleton peak torque increased from low to high with the motors as designed and the spring peak was similar to the medium motor (Figure 8b). The negative biological ankle power was reduced for the spring and motor conditions, but the late stance peak was only reduced for the spring compared to exoskeleton no assistance (Figure 8d). The exoskeleton positive power peak increased from low to high motor, but the spring peaked similar to motor low. The exoskeleton assistance conditions increased plantarflexion in late stance (Figure 8e).

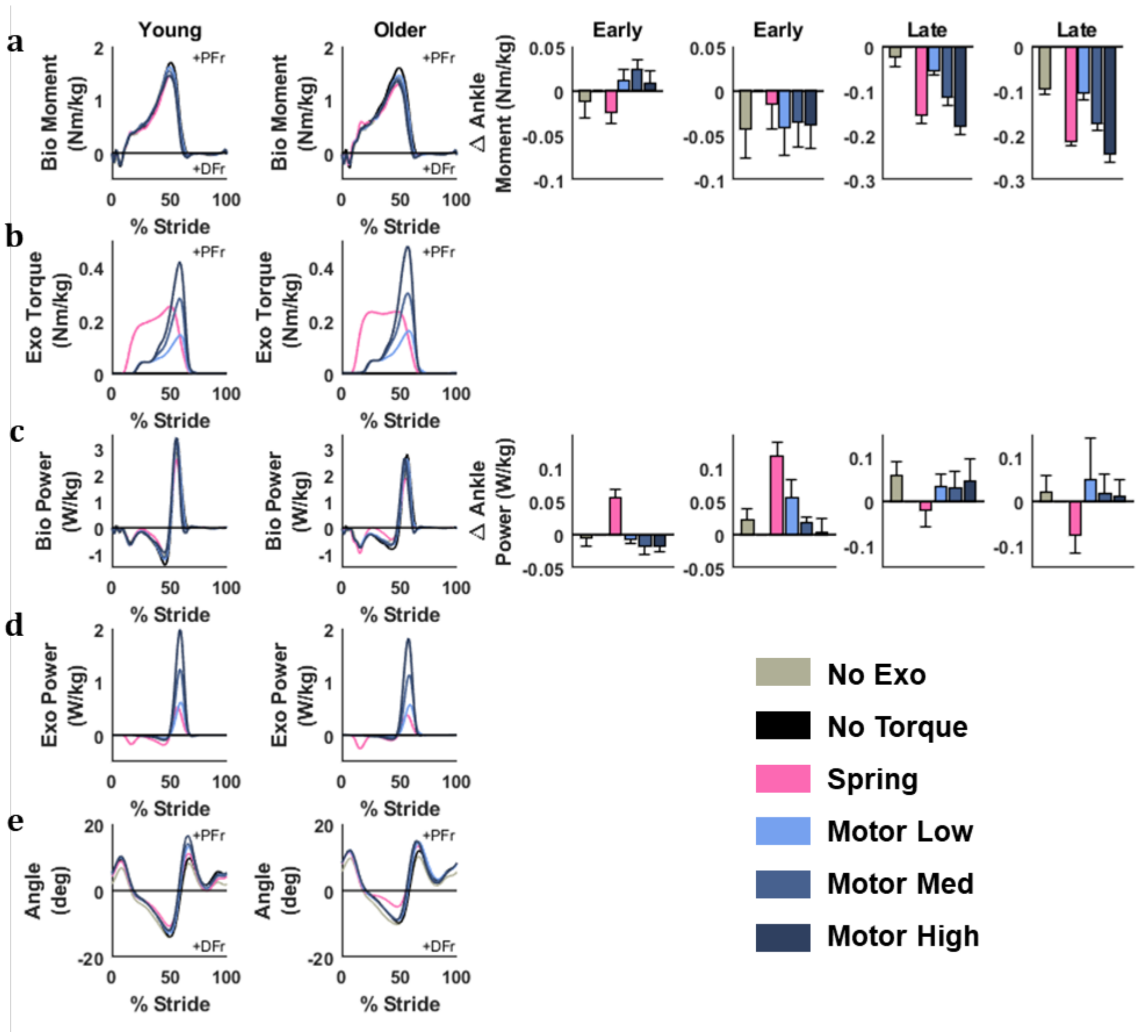

Figure 8 - Ankle and Exoskeleton Dynamics and Kinematics: The average a) biological contribution to the ankle moment, b) the exoskeleton torque, c) the biological contribution to the ankle power, d) the exoskeleton power, and e) the ankle angle for young (left) and older (right) adults. The bar graphs are the corresponding average change from the exoskeleton no assistance condition over early (10-40%) and late (40-70%) stance for young (left) and older (right) adults.

When comparing young to older adult responses, the main difference came from the ankle angle (Figure 8e). Young adults were more plantarflexed throughout late stance with a larger peak plantarflexion than exoskeleton no assistance. Older adults had only a

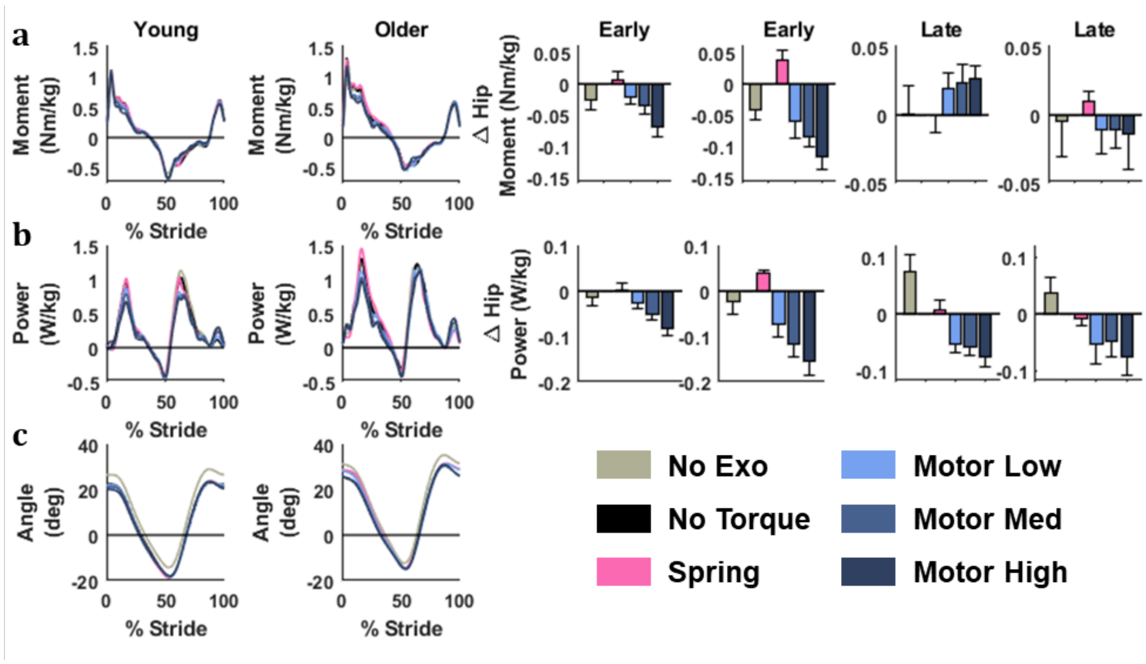

Figure 9 - Hip Dynamics and Kinematics: The average a) biological contribution to the hip moment, b) the biological contribution to the hip power, and c) the hip angle for young (left) and older (right) adults. The bar graphs are the corresponding average change from the exoskeleton no assistance condition over early (10-40%) and late (40-70%) stance for young (left) and older (right) adults.

small increase in peak plantarflexion with the motor exoskeletons. The spring assistance reduced the dorsiflexion angle from 20-60% of the stride.

3.3.3 Hip Dynamics

Both age groups reduced hip extensor moment in early stance and hip flexor moment at push off with motor assistance (Figure 9a). Hip power was also reduced in early

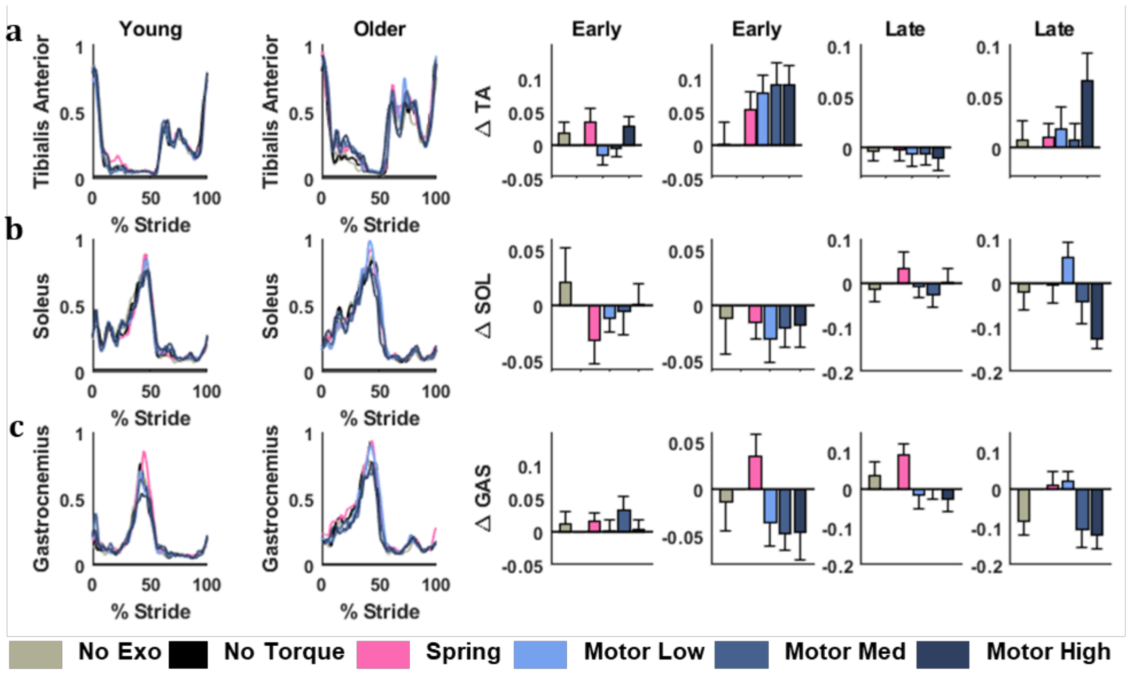

Figure 10 - Ankle EMG: Average a) tibialis anterior (TA), b) soleus (SOL), and c) lateral gastrocnemius (GAS) activation across a stride for young (left) and older (right) adults. The bar graphs are the corresponding average change from the exoskeleton no assistance condition over early (10-40%) and late (40-70%) stance for young (left) and older (right) adults.

stance for both age groups (Figure 9b). Hip moment and power for spring assistance was similar to no assistance. Hip angle was also similar across exoskeleton conditions (Figure 9c). At late stance young and older adults deviated. Young adults reduced hip flexion moment and peak power with the motor conditions to a greater extent than the older adults.

Averages were calculated to compare early versus late stance changes because the spring assistance acts over stance and the motor acts over late stance (Figure 6b). The spring reduced the ankle moment across stance, increased ankle power in early stance and decreased ankle power in late stance for both groups. The motor conditions reduced ankle moment in late stance, decreased hip moment and power in early stance, and decreased hip

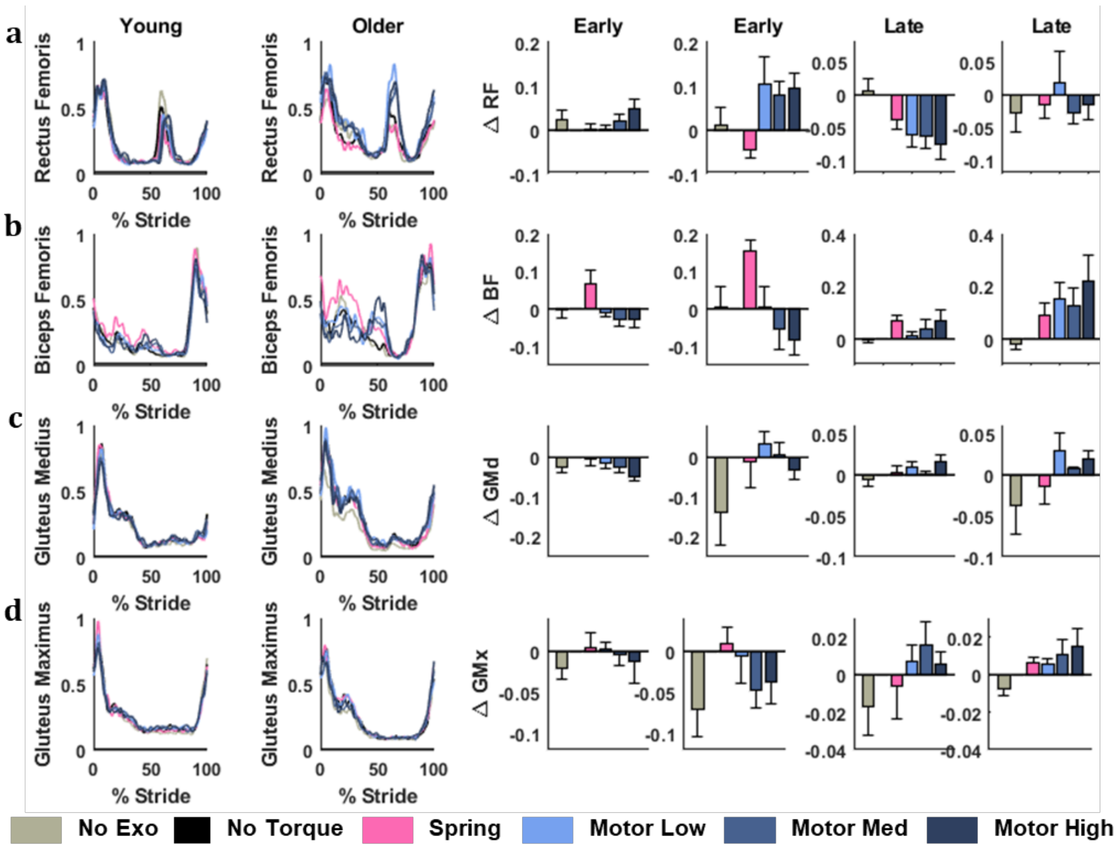

Figure 11 - Hip EMG: Average a) rectus femoris (RF), b) biceps femoris (BF), c) gluteus medius (GMd), and d) gluteus maximus (GMx) activation across a stride for young (left) and older (right) adults. The bar graphs are the corresponding average change from the exoskeleton no assistance condition over early (10-40%) and late (40-70%) stance for young (left) and older (right) adults.

power in late stance for both groups compared to exoskeleton no assistance. The age groups deviated in strategy at the hip for the spring and in early stance at the ankle for the motors. The older adults in springlike assistance had a greater reduction in ankle power during late stance, increased the hip moment across stance, and increased hip power in early stance compared to young adults. The older adults with motor assistance reduced ankle moment and increased ankle power in early stance, and reduced hip moment in late stance.

3.3.4 Electromyography (EMG)

To dive deeper into the muscles contributing to joint level changes, I measured EMG of the ankle and hip musculature. The EMG was again split into early and late stance averages to see how the exoskeletons responded at different times across stance. Both groups increased tibialis anterior (TA) and gastrocnemius (GAS) and reduced soleus (SOL) activity in early stance with spring assistance (Figure 10). For the motor assistance, both groups reduced SOL activity across stance and GAS activity in late stance. Older adults differed from young with much lower activity across all calf muscles in late stance walking in spring assistance. With the motor assistance, they had increases in TA activity across stance with greater increases in early stance and decreased SOL and GAS activity across stance more than young adults.

At the hip, both age groups reduced rectus femoris (RF) activity in late stance, increased biceps femoris (BF) activity across stance, and increased then decreased gluteus maximus (GMx) activity across stance with spring assistance (Figure 11). For the motor, both groups increased and then decreased RF activity and decreased then increased BF, gluteus medius (GMd), and GMx activity across stance. Older adults deviated minimally for hip muscle activity from young adults with exoskeleton assistance.

3.3.5 Power

It has been shown that exoskeleton work, not torque, leads to greater reductions in metabolic cost [67]. To confirm that in my data, ankle, hip, and exoskeleton positive power over a stride were compared to the change in net metabolic power for the no assistance condition with both age groups combined. Increases in ankle and exoskeleton positive power coincided with a reduction in hip positive power and net metabolic power (Figure 12).

3.4 Discussion

The purpose of this study was to explore ankle exoskeleton assistance strategies to reduce metabolic cost across the lifespan. Based on previous studies that focused predominantly on young adults, we expected motor-like assistance to outperform spring-like in reducing net metabolic cost. Our results agree with this hypothesis and found the motor-like assistance outperformed the spring-like assistance (Figure 7). This result is consistent with previous literature that showed exoskeleton profiles that do more positive work are more effective at reducing energy. For example, in a study comparing assistance profiles increasing work or torque, high work profiles reduced metabolic cost by 17% while high torque profiles increased it [67]. Additionally, when using human-in-the-loop optimization to reduce metabolic cost, the assistance generated created a large amount of positive power at push off. This may reflect lengthening muscle fascicles and slowing the shortening velocities in late stance to lower metabolic cost [23].

In agreement with our second hypothesis, the older adults benefited more from ankle exoskeleton assistance, at least for the motors (Figure 7, Table 1). Ankle exoskeleton assistance was expected to offset the reliance on the less efficient hip musculature seen with aging. The highest motor provided 23.3% reductions for OA compared to 16% for YA while the spring-like assistance was worse for OA (Figure 7). The older adults were

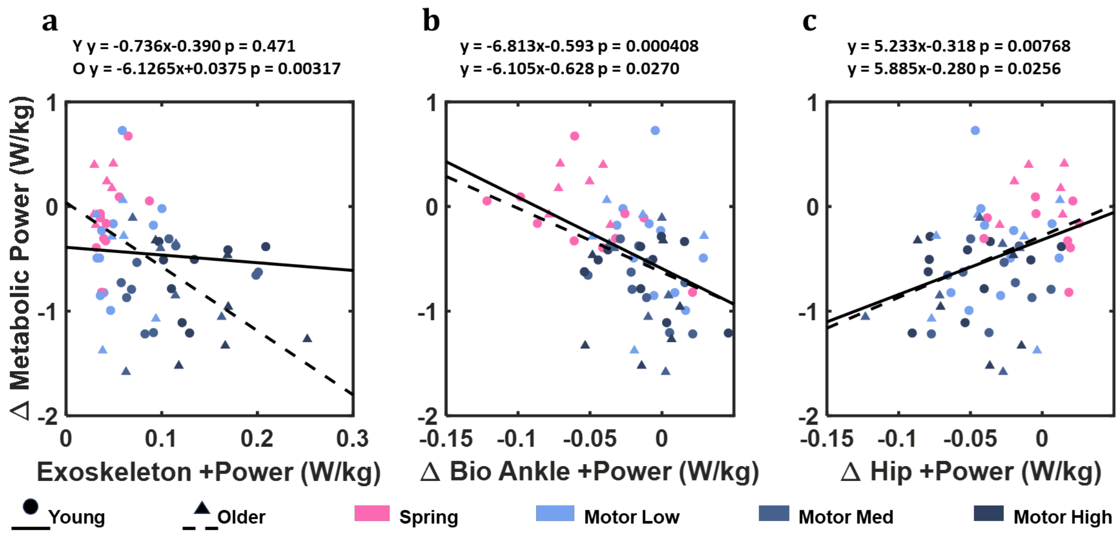

Figure 12 - Metabolic and Mechanical Power: The relationship between the change in net metabolic power from exoskeleton no assistance with a) exoskeleton, b) ankle, and c) hip biological positive power for young (circle, solid) and older (triangle, dashed) adults. A reduction in metabolic power coincided with an increase in ankle positive power and a reduction in hip positive power for young adults. A reduction in metabolic power coincided with an increase in exoskeleton and ankle positive power and a reduction in hip positive power for older adults.

able to receive a greater bang per buck with the motor-like exoskeleton assistance consistent with our theory to load the ankle more to take advantage of the higher efficiency at the ankle due to the Achilles tendon [11]. This is further corroborated in Figure 12 showing OA reduced their metabolic cost more per unit of exoskeleton positive power (OA slope = -6.13 W/kg, YA slope = -0.736 W/kg). Interestingly, the sweet spot in motor-like assistance for the young adults was with a peak torque of 20 Nm and 30 Nm for older (Figure 7). This is consistent with literature in simulations and experiments that show there is a sweet spot of timing and magnitude when reducing metabolic cost with exoskeletons [68]–[71].

Interestingly, both young and older adult users utilized the ankle motor-like assistance to predominantly off load the hip, rather than the ankle (Figure 8-12). The muscle activation follows similar patterns to those of the kinetics at the ankle and hip as hypothesized in Figure 6c (Figure 10-11). In late stance, exoskeleton assistance offloads the plantar flexors, the gastrocnemius and soleus, a small amount (Figure 10). At the same time, the hip flexors are offloaded as seen by the rectus femoris (Figure 11). Then in early stance when the contralateral limb is receiving assistance, the hip extensors, biceps femoris, gluteus medius, and gluteus maximus are offloaded. This has been seen before in other ankle exoskeletons but not all comparable exoskeletons [72], [73].

This work has many implications for assistive technology for older adults. For example, the older adults had a higher added mass penalty so lighter devices are more important. Until lighter motorized devices can be made, passive devices with springs can be investigated further for older adults. Passive devices have the advantage of being lightweight. However, for this study, the stiffness and timing were not optimized for older adults potentially misrepresenting what passive devices can do for older adults. If optimization is not enough, more creative systems may be necessary to create torque profiles from springs that are more impulsive by using nonlinear springs, complex clutches, or a pseudoactive device.

Table 1 – Metabolic Power: Average net metabolic power for all participants and group averages for young and older adults.

Pmet (W/kg)	No Exo	No Torq	Spring	Motor Low	Motor Med	Motor High
YA04	3.34	4.35	4.24	3.53	3.13	3.14
YA06	2.84	3.42	2.60	2.93	2.69	3.09
YA07	3.70	4.02	4.70	3.86	3.65	3.64
YA08	2.75	3.38	3.06	2.89	2.59	2.87
YA09	3.25	3.54	3.24	2.55	2.33	3.04
YA10	2.22	2.84	2.89	2.82	2.18	2.21
YA11	2.80	3.42	3.35	3.19	2.89	2.63
YA13	3.79	4.47	4.08	3.62	3.60	3.36
YA14	4.08	4.49	4.58	5.21	4.18	4.07
YA16	4.27	4.65	4.49	4.47	4.02	4.36
Young	3.30	3.86	3.72	3.51	3.13	3.24
OA04	3.48	4.85	4.77	3.47	3.27	3.33
OA06	3.83	4.44	4.85	4.50	4.08	3.11
OA08	3.78	4.63	4.81	4.34	4.23	4.17
OA10	3.48	4.03	4.27	3.75	3.18	3.07
OA11	3.11	4.04	3.87	2.97	2.99	2.78
OA12	3.11	3.09	3.49	3.02	2.98	2.77
Older	3.46	4.18	4.34	3.68	3.46	3.20

CHAPTER 4. LINKING CALF MUSCLE-TENDON PROPERTIES TO USER PERFORMANCE IN ACTIVE AND PASSIVE ANKLE EXOSKELETONS

4.1 Introduction

Exoskeletons have advanced over the last ten years to reduce metabolic cost up to 50% when walking [19], [63]. There have been multiple types of exoskeletons created to reduce metabolic cost including passive, active, and pseudo-passive. However, they all set out to produce a desired torque profile. The passive device creates a torque profile using a spring and clutch to store and release energy over stance to provide a spring in your step. The active devices create an impulsive torque profile using a motor to provide additional push off in late stance. Pseudo-passive combines active and passive components to create a combination profile. What we have found over the years is that the magnitude and timing is critical for creating positive power to reduce metabolic cost [68], [70], [71]. When comparing ankle exoskeleton profiles increasing work or torque, the exoskeletons with more work and positive power, lead to greater reductions in metabolic cost [67]. When using human-in-the loop optimization, with ankle exoskeletons, to reduce metabolic cost the profile that emerged was an impulsive torque with high positive power in late stance [24]. However, it is unclear why these shapes are successful, especially at the physiological level.

The ankle plantar flexors have a particularly unique morphology consisting of muscles with shorter fascicles and long elastic tendons. The Achilles tendon allows the muscle to decouple from the joint to produce force more economically and thus reduce metabolic cost. There is some work showing that the stiffness of the tendon is optimally tuned for the metabolic cost of walking [74]. There are many conditions/instances that influence the stiffness of the Achilles tendon including working out, diabetes, aging, and cerebral palsy [14], [17], [75]–[77]. Combined, the stiffness of the Achilles tendon can affect the metabolic cost of walking. For example, using the equation in Figure 13e, if the tendon stiffness is reduced, the tendon length will increase. The muscles will operate at shorter, less optimal lengths and higher activations raising metabolic cost (Figure 13d). However, it is unknown when combining differences in Achilles tendon stiffness with ankle exoskeleton assistance, how the two will interact to modify metabolic cost.

While the tendon properties influence metabolic cost, it is likely the muscle properties do as well. While the exoskeletons are placed on joints, it is thought they operate by modifying muscle contractile dynamics to drive metabolic cost. In young adults using spring-like assistance, exoskeletons reduce the mechanical demand at the ankle, lengthen muscles in early stance, and reduce muscle activation to lower metabolic cost [78], [79]. While muscle level changes have not been recorded in vivo with motor-like assistance, data driven simulations found similar results as spring-like but with the changes primarily occurring in late stance [80]. Another example, using the equation in Figure 13f, if the exoskeleton offloads the force required to walk, the tendon length will decrease. The muscle will operate at longer, more optimal lengths, and lower activations reducing metabolic cost (Figure 13d). Understanding how users' muscle and tendon morphology interact with an exoskeleton's torque profile is still unclear.

The purpose of this study was to evaluate the calf muscles and tendon's role in modifying metabolic cost during walking in springlike and motorlike ankle exoskeleton assistance profiles. Participants walked in commercially available ankle exoskeletons (Dephy Inc, Boxborough, MA) with springlike and motorlike assistance profiles while measuring muscle and tendon properties with ultrasound. Based on experimental and modeling literature, we hypothesize, the exoskeletons would lower muscle force and activation, lengthen muscle fascicles to reduce metabolic cost across 1a) stance for the springlike and 1b) late stance for motorlike. Secondarily, we hypothesize that lower Achilles tendon stiffness would lead to a larger reduction in metabolic cost from the exoskeleton assistance. Using Figure 13g to explain, the force required to walk in the exoskeleton will be less than without the exoskeleton, creating a double negative change in force, or positive numerator. Then with lower tendon stiffness we see a larger shift toward longer lengths. It has been shown that operating at shorter lengths will raise your metabolic cost, so longer lengths should reduce metabolic cost [32]. Our anticipated results are intended to provide valuable insight into how ankle exoskeletons interact with the underlying physiology linking structure to function.

4.2 Methods

4.2.1 Participants

Nine healthy younger adults (2 female, 7 male; age: 21.11 ± 1.96 years; height: 176.72 ± 10.72 cm; mass: 74.61 ± 17.19 kg) and four healthy older adults (4 female; age: 72.5 ± 5.80 years; height: 165.11 ± 2.25 cm; mass: 59.54 ± 4.51 kg) participated in the

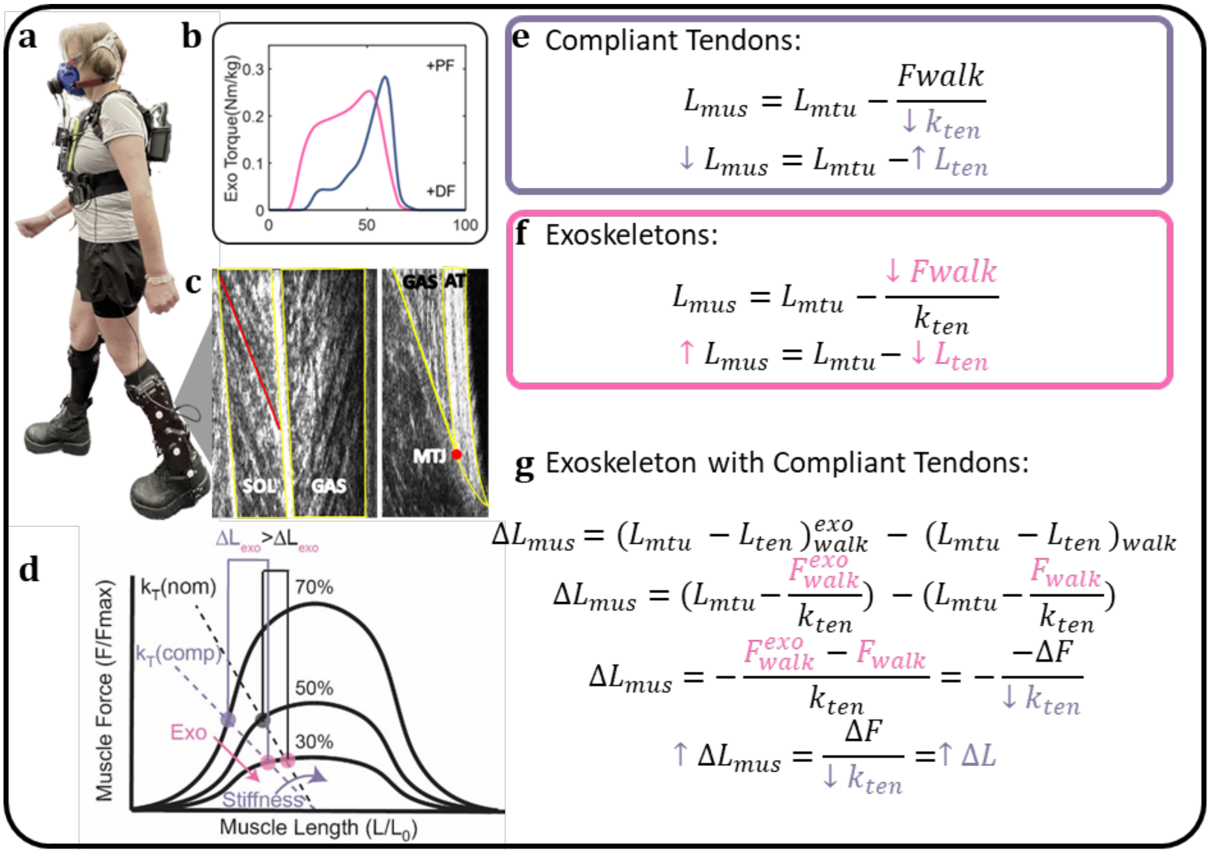

Figure 13 - a) Participant walking with ankle exoskeletons and portable metabolic system. b) The average exoskeleton torque provided over a stride. c) Soleus fascicle from B-mode ultrasound recorded over the medial gastrocnemius (left) and the gastrocnemius-Achilles tendon muscle tendon junction (MTJ) (right) for calculating stiffness. d) Hypothesized mechanism for reduced tendon stiffness (purple) and ankle exoskeleton assistance (pink) on theoretical force-length curves at different activations (30, 50, 70%). e) Mathematical hypothesis for how complaint tendons would modify operating ranges on force-length curves when walking. f) Mathematical hypothesis for how exoskeleton assistance would modify operating ranges on force-length curves when walking. g) Mathematical hypothesis for how people with complaint tendons would benefit more from exoskeleton assistance by modifying the operating ranges on force-length curves when walking.

study. The participants met the inclusion criteria of not having a history of musculoskeletal injuries or vestibular impairments, and the ability to walk for a minimum of 60 minutes within a 90-minute time frame. All participants signed an informed consent approved by the Institutional Review Board at Georgia Institute of Technology.

4.2.2 Exoskeleton

Commercially available bilateral ankle exoskeletons (Dephy Inc, Boxborough, MA) with plantarflexion assistance were used with custom controllers (Figure 13a). The exoskeletons have on-board encoders and inertial measurement units for state-based control. Participants walked in three conditions: (1) exoskeletons without assistance, (2) exoskeletons with springlike assistance, and (3) exoskeletons with motorized assistance strategies with a peak torque of 20 Nm. The springlike assistance used impedance control without damping to apply torque with a spring stiffness of 70 Nm/rad [23]. The assistance began at flat foot after heel strike and ended at toe off mimicking a spring and clutch mechanism to time when to release the stored energy[22], [23]. The torque profile for the motor assistance was a human-in-the-loop optimized spline generated from four nodes: peak time, rise time, fall time, and peak torque (Figure 13b) [24]. The parameters used were optimized for metabolic cost with the average younger adults' parameters from Zhang et al. The only difference was peak torque was preset to 10, 20, and 30 Nm for this experiment.

4.2.3 Protocol

Participants completed testing over one dynamometer session and three walking sessions. Participants walked at 1.25 m/s on an instrumented split-belt treadmill (CAREN, Motek, Netherlands). The four testing sessions were 1) muscle-tendon characteristics, 2) habituation, 3) gait mechanics, and 4) steady-state metabolic energy consumption. Waiting periods between sessions were minimized to allow for learning and retention (2-7 days on average).

4.2.4 Muscle-Tendon Characteristics

Force-length curves at 100, 66, 33, and 0% activation were measured for each participant laying supine with their right knee flexed 120° to minimize the contribution of the gastrocnemius [42]. All trials took place on a dynamometer (Biodex Medical Systems, Shirley, NY) to measure the torque. Torque feedback from the previous trial and verbal encouragement were used to reach maximal activation. EMG feedback was used to reach 33 & 66% activation of the soleus. Passive trials were ensured to have no activation of the surrounding muscles. Five angles (20PF, 10 PF, 0, 15 DF, and max DF) were chosen so fascicle lengths spanned the greatest portion of the force-length curve.

Ultrasound was placed over the medial gastrocnemius to image the soleus (Figure 13c). Motion capture was used to calculate moment arms and ensure minimal lift of the heel during contractions (markers: medial epicondyle, medial malleolus, head of the 1^{st} metatarsal, dynamometer axis of rotation, dynamometer offset, and on the pedal at half the length of the foot). Muscle activity was recorded using surface EMG of the soleus.

We collected motion capture data (200 Hz) (ViconMotion Systems, UK), torque on a dynamometer (1000 Hz) (Biodex Medical Systems Inc., USA), ultrasound (100 Hz), and EMG (1000 Hz) at the corresponding sampling frequencies. The EMG data was cleaned by removing the mean, bandpass filtered (20-450 Hz), notch filtered (60 Hz), absolute value rectified, and lowpass filtered at 6 Hz. Kinematics and kinetics were filtered at 6 Hz. Soleus fascicle lengths were measured using Ultratrack [46].

Achilles tendon stiffness measurements were calculated from plantarflexion contractions with a straight leg on the dynamometer (Biodex Medical Systems Inc., USA). The torque from the dynamometer was converted to force using the Achilles tendon moment arm. A b-mode ultrasound probe (Telemed, Vilnius, Lituani) was placed over the muscle tendon junction (MTJ) of the gastrocnemius and the Achilles tendon. Participants were asked to slowly contract as hard as possible. The data was parsed into 10% force intervals to track the length change in ImageJ (Bethesda, MD). A linear regression was fit to the force and length data in the 50-100% range of force output [49], [50]. The slope of the line is the stiffness in N/mm.

4.2.5 Habituation

Previous work has shown the importance of habituation for the acceptance of assistance from exoskeletons [65]. In the first session, each participant walked with spring-like assistance for three, seven-minute intervals for a total of 21 minutes. They also walked with motorized assistance for three, seven-minute intervals while increasing peak torque (10, 20, 30 Nm) with each bout.

4.2.6 Gait Mechanics

The primary goal for the second session was to obtain biomechanical measurements. We instrumented participants with surface electromyography to record muscle activity on the right soleus (SOL) and ultrasound over the medial soleus to get fascicle length and velocity. We placed reflective markers on the lower limbs and torso to collect motion capture. Participants were rehabituated to the exoskeleton assistance for 6-minutes prior to walking with exoskeletons without assistance, with springlike assistance, and with motorized assistance. After each condition's habituation trial, participants walked for 2 minutes. The exoskeleton assistance order was randomized for each participant.

4.2.7 Steady-State Metabolic Energy Consumption

Participants began the third session with a five-minute standing trial to record their metabolic resting baseline. Participants were fitted to a portable indirect calorimetry device (COSMED K5, Rome, Italy) to collect energy expenditure for each condition (Figure 13a). We repeated the habituation process conducted on the gait mechanics session. The participants walked for six minutes at each of the three conditions in a randomized order with the respective habituation trial before.

4.2.8 Biomechanics Measurements

Kinematics were measured using a 10-camera motion capture system (100 Hz, Vicon, Oxford, UK) and a 44 marker lower limb and torso marker set. Ground reaction forces were captured with an instrumented split-belt treadmill (2000 Hz, CAREN, Motek, Netherlands). Inverse kinematics and inverse dynamics were calculated using OpenSim v4.0 (generic model 2392 with scapula) [51]. Joint angle data was filtered at 6 Hz (4^{th} order zero-phase Butterworth) and ground reaction forces were filtered at 15 Hz (4^{th} order zero-phase Butterworth) prior to inverse dynamics calculations. The biological contribution to the total ankle joint moment was calculated by subtracting the estimated exoskeleton torque from the total ankle moment. Joint angles, moments, and powers were calculated for the sagittal plane at the ankle. All data was separated by stride using a 30 N threshold of the vertical ground reaction force. The stride average is the average of five strides. Joint moments and powers were integrated over early (0-40 % stride) and late stance (40-70% stride) for each individual and then group averaged.

Muscle activity of the soleus was measured using surface EMG on the right leg (2000 Hz, Avantis, Delsys, Natick, MA). The EMG locations were shaved and exfoliated for improved signal. The data was processed by removing the mean, 20-400 Hz bandpass filtered (8^{th} order zero-phase butterworth), rectified using the absolute value, and smoothed using a 20 Hz lowpass filter (3^{rd} order zero-phase butterworth). The EMG amplitude was normalized to the peak from the filtered stride averaged EMG from the exoskeleton condition without assistance for each participant. The integrated EMG was computed as the time-integral of the filtered EMG averaged across each stride divided by the time period over which it was integrated for early stance (0-40%), and late stance (41-70%).

Muscle fascicles of the medial soleus were collected using B-mode cine ultrasound (ArtUS EXT-1H acquisition unit, 60 mm long LV8-5N60-A2 probe, Telemed, Vilnius, LT). The probe was placed over the belly of the medial gastrocnemius where the fascicles were easiest to visualize (50-117 Hz). Fascicles were tracked in Ultratrack to measure length and pennation angle [46]. Fascicle velocity was calculated as the derivative of the fascicle length over time.

4.2.9 Metabolic Cost Measurements

Indirect calorimetry was collected using a portable system (COSMED K5, Rome, Italy) as breath-by-breath measurements for each walking trial (Figure 13a). Each trial lasted six minutes to ensure steady-state metabolic rate was achieved. Metabolic power was calculated using Brockway's equation and averaging the last minute of the trial [66]. Net metabolic power was calculated by subtracting the standing trial average from the walking trial averages. The exoskeleton no assistance trial was then subtracted off to get the change and percent change in net metabolic power.

4.2.10 Statistics

Statistical analysis was performed in SPSS (IBM SPSS Statistics, Armonk, NY) with Laerd Statistics tutorials and guides. For hypotheses 1a and 1b, one-way repeated measures analysis of variance (ANOVA) with Shapiro-Wilk test of normality was used to find main effect of exoskeleton assistance in early or late stance on muscle force, activation, fascicle length, and metabolic cost. Early stance fascicle velocity, early stance force, and soleus activation in late stance for the exoskeleton without assistance violated the normality test ($p > 0.05$). However, ANOVA was still used on this data because it is robust to deviations of normality. On the significant main effects, a post hoc pairwise comparison with Bonferroni correction was used to test which exoskeleton conditions were different for a given variable. A linear regression was used to compute correlation coefficients between metabolic cost and the muscle-tendon parameters for the secondary hypothesis: tendon stiffness, muscle strain, maximum force, and optimal length.

4.3 Results

All data reported will be mean ± standard deviation. A significant difference in metabolic cost for the springlike assistance was not found ($p > 0.05$) (Figure 14). Springlike assistance did reduce fascicle force in late stance (NoTorq = 15.99 ± 3.19 N/kg, Spring = 12.60 ± 3.13 N/kg, $p < 0.001$) (Figure 16a). Fascicle power, soleus activation, fascicle length, and fascicle velocity were not found to be significantly different from the exoskeleton no torque condition ($p > 0.05$) (Figure 16 b-e). The change in soleus activation from the exoskeleton no assistance condition was found to be significantly higher than for motorlike assistance in late stance (Spring = 0.53 ± 0.11, Motor = 0.38 ± 0.12, $p = 0.015$) (Figure 16c).

Motorlike assistance lowered metabolic cost significantly ($p < 0.05$) (Figure 14). Motorlike assistance reduced fascicle force and soleus activation in late stance (Force: NoTorq = 15.99 ± 3.19 N/kg, Motor = 13.43 ± 2.60 N/kg, $p < 0.001$; Activation: NoTorq = 0.52 ± 0.083, Motor = 0.38 ± 0.12, $p = 0.014$) (Figure 16a, c). Fascicle power, fascicle length, and fascicle velocity were not found to be significantly different from the exoskeleton no torque condition ($p > 0.05$) (Figure 16b, d, e).

A linear regression was used to test hypothesis 2: people with lower Achilles tendon stiffness would benefit more from exoskeleton assistance. A significant correlation was not found ($p > 0.05$) (Figure 18a). Additionally, a correlation was not found between the change in net metabolic power and any muscle properties ($p > 0.05$) (Figure 18b-d).

4.4 Discussion

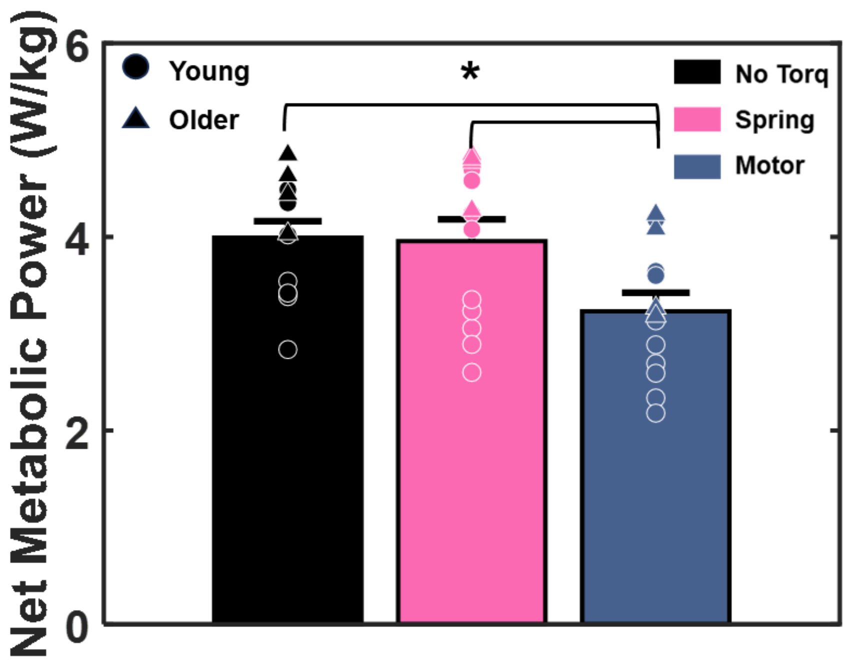

Figure 14 - Average net metabolic power while walking in the exoskeleton without assistance (black), springlike assistance (pink), and motorlike assistance (blue). The data points represent participants with the circles being young adults and triangles being older adults. The motor was statistically different from both no torque and springlike assistance ($p < 0.05$).

The purpose of this study was to evaluate the calf muscle and tendon's role in modifying metabolic cost during walking in springlike and motorlike ankle exoskeleton assistance profiles. We used the Dephy ankle exoskeleton to provide spring and motorlike torque profiles while measuring muscle and tendon morphology using ultrasound (Figure 15a, c). As expected, the springlike assistance provided a large torque across stance that reduced the biological ankle moment and power at push off (Figure 15). The motorlike assistance provided an impulsive torque in late stance reducing biological ankle moment but not power at push off (Figure 15). In line with previous literature, we found motor

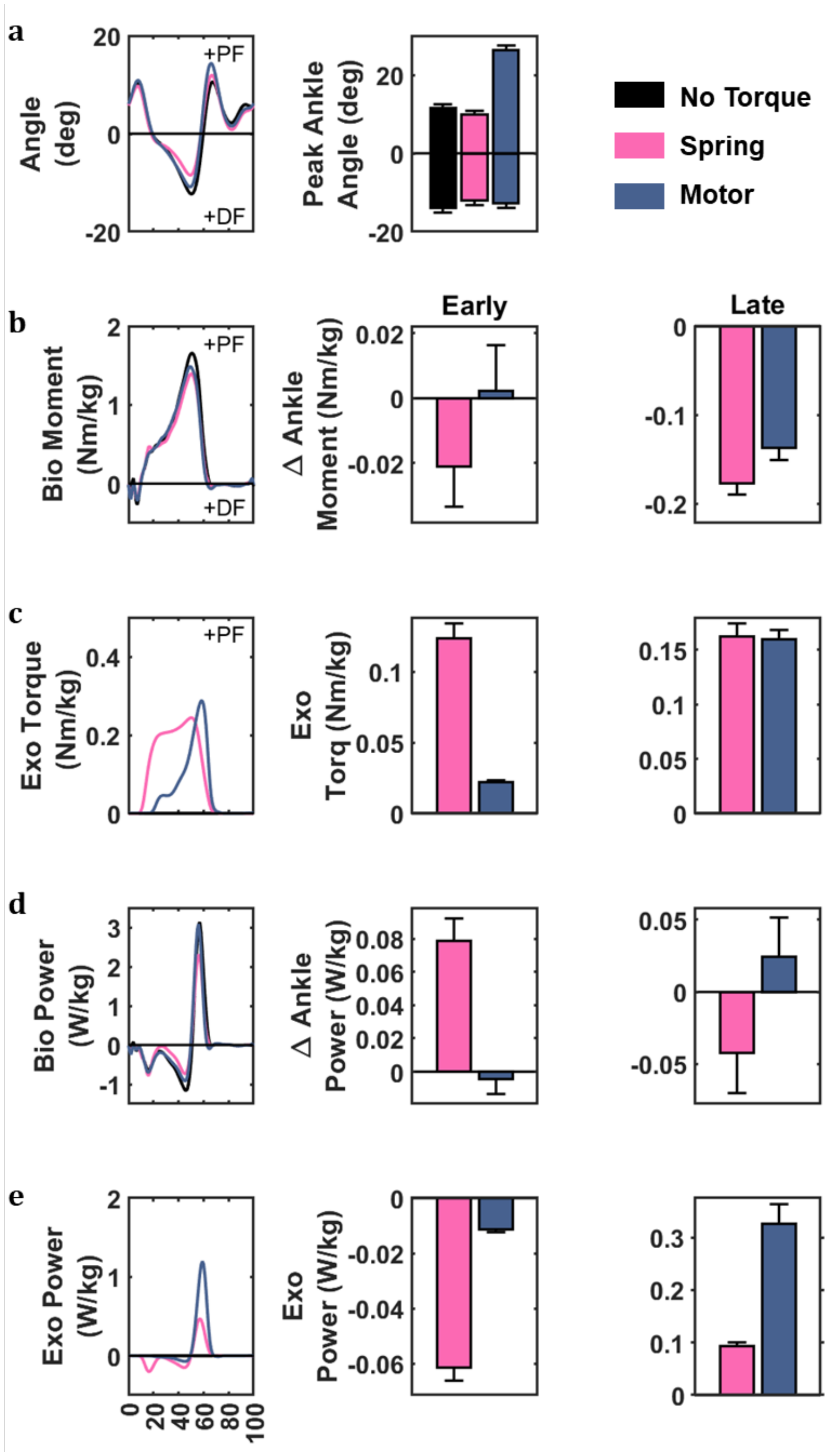

Figure 15 - Ankle Inverse Kinematics and Dynamics: The stride average a) ankle angle, b) biological contribution to the ankle moment, c) exoskeleton torque for all exoskeleton conditions, d) biological contribution to the ankle power, and e) the exoskeleton power with early (0-40% stride) and late stance (40-70% stride) averages.

strategies were more effective than springlike at reducing metabolic cost from the no torque condition (Figure 14) [19], [21]–[24], [63]. However, our main hypotheses were based on how the muscle and tendon morphology would modify metabolic cost when walking in spring and motorlike exoskeleton assistance.

Based on literature from young adults, we hypothesized, springlike exoskeleton assistance would lower muscle force and activation, lengthening muscle fascicles to reduce metabolic cost across stance. Surprisingly we found no change in metabolic cost from the exoskeleton no torque condition ($p < 0.05$) (Figure 14). This is different from literature where a small reduction in metabolic cost, 4-7% was found [22], [23] This could mean the stiffness used was not the optimal stiffness for this cohort. We found partial agreement with the rest of the hypothesis; the fascicle force was lower than the no torque condition but only during late stance Figure 16a. This is consistent with previous literature for young adults [79]. Fascicle power, muscle activation, fascicle length, and fascicle velocity were not statistically different from those found in the no torque condition in early or late stance (Figure 16). This is not in agreement with literature. In a couple studies muscle activation was reduced [78], [79]. However, only one study to date has measured muscle fascicle data during exoskeleton assistance [79]. This study found a reduction in muscle fascicle forces, activation, increase in fascicle length, and shortening velocity. One difference between our study and theirs is they tested multiple spring stiffnesses and found statistical significance for activation, length, and velocity only at stiffnesses higher than what was used in this study. Additionally, since we didn't see metabolic changes with the spring it would make sense, we wouldn't see changes in other measures as well.

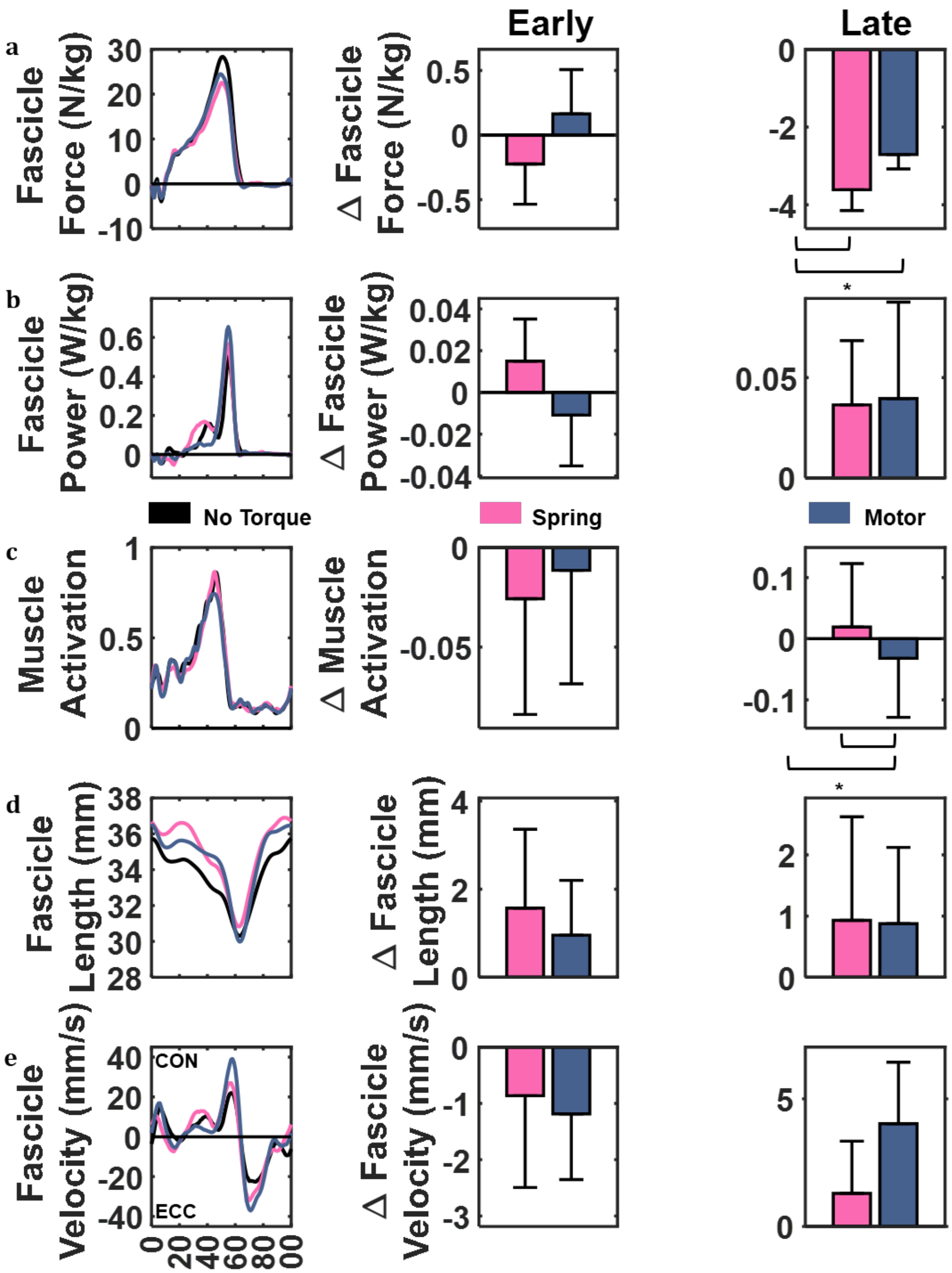

Figure 16 - Soleus Muscle Tendon Dynamics: The stride average soleus a) fascicle force, b) fascicle power, c) activation, d) fascicle length, and e) fascicle velocity with early (0-40% stride) and late stance (40-70% stride) averages. Both fascicle force and soleus activation were statistically different in late stance (* $p < 0.05$).

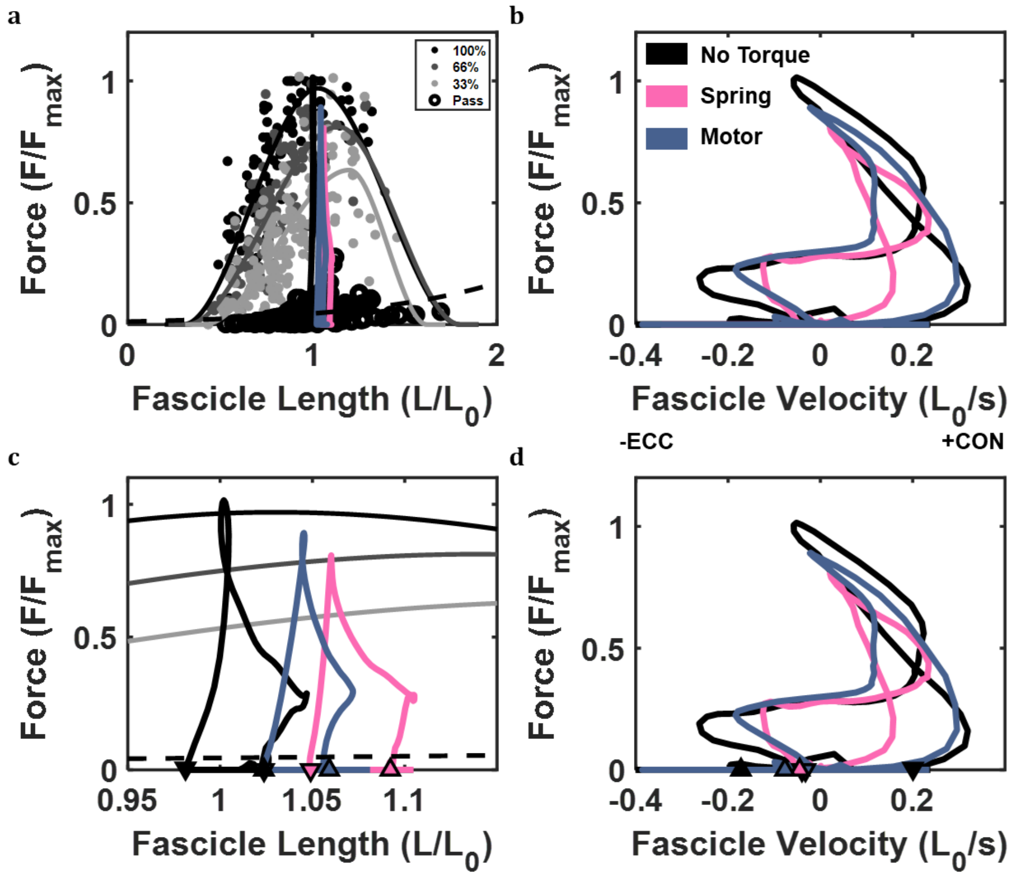

Figure 17 - Force-Length and -Velocity Curves: a) group average 0, 33, 66, and 100% activation force-length curves and the group average operating range while wearing exoskeletons with no assistance, springlike assistance, and motorlike assistance. c) figure a zoomed in to see the workloops. Up triangles represent heel strike and down triangles represent toe off. b) Normalized force-velocity curve with group average operating range while wearing exoskeletons with no assistance, springlike assistance, and motorlike assistance. d) figure b zoomed in to see the workloops. Up triangles represent heel strike and down triangles represent toe off.

Our corollary hypothesis was motorlike assistance with ankle exoskeletons would lower muscle force and activation, lengthen muscle fascicles in late stance to reduce metabolic cost. The hypothesis was based on data driven simulations in young adults wearing an ankle exoskeleton with a late stance impulsive torque. They predicted the metabolic reduction came from lowering the fascicle force and activation and operating at longer, more efficient lengths, primarily in late stance [80]. Consistent with previous literature, we found a reduction of 18.12% in metabolic cost while walking with motorlike assistance [21], [24], [63] (Figure 14). Consistent with the data driven simulation, we found a reduction in fascicle force and soleus activation in late stance (Figure 16) [80]. Inconsistent with our hypothesis and the simulations, no statistically different changes were seen for fascicle length, velocity, or power. Thre are a couple reasons we may not be seeing changes in length, velocity, or power (Figure 16). First off, the effect sizes for these variables are very small <0.1. Secondly, it could be small changes in multiple variables that lead to the reduction in metabolic cost. For example, the fascicle lengths were trending towards longer in early stance while maintaining a short length at push off. This increased the velocity of the fascicle which increased the peak power while generating a lower force at push off. Combined, they're creating more power with less force and activation, reducing metabolic cost. Lastly, in our previous work we found most people maintained their ankle contribution while offloading the hip. This means the changes at the muscle level would be occurring in the hip musculature, not the ankle.

Based on longer lengths leading to less metabolically costly contractions, we hypothesized that lower Achilles tendon stiffness would lead to a larger reduction in metabolic cost. To recap, adding an exoskeleton will result in positive change in force and when divided by a smaller tendon stiffness creates a larger shift toward longer lengths (Figure 13d). Our hypothesis was not confirmed, we saw no correlation between metabolic

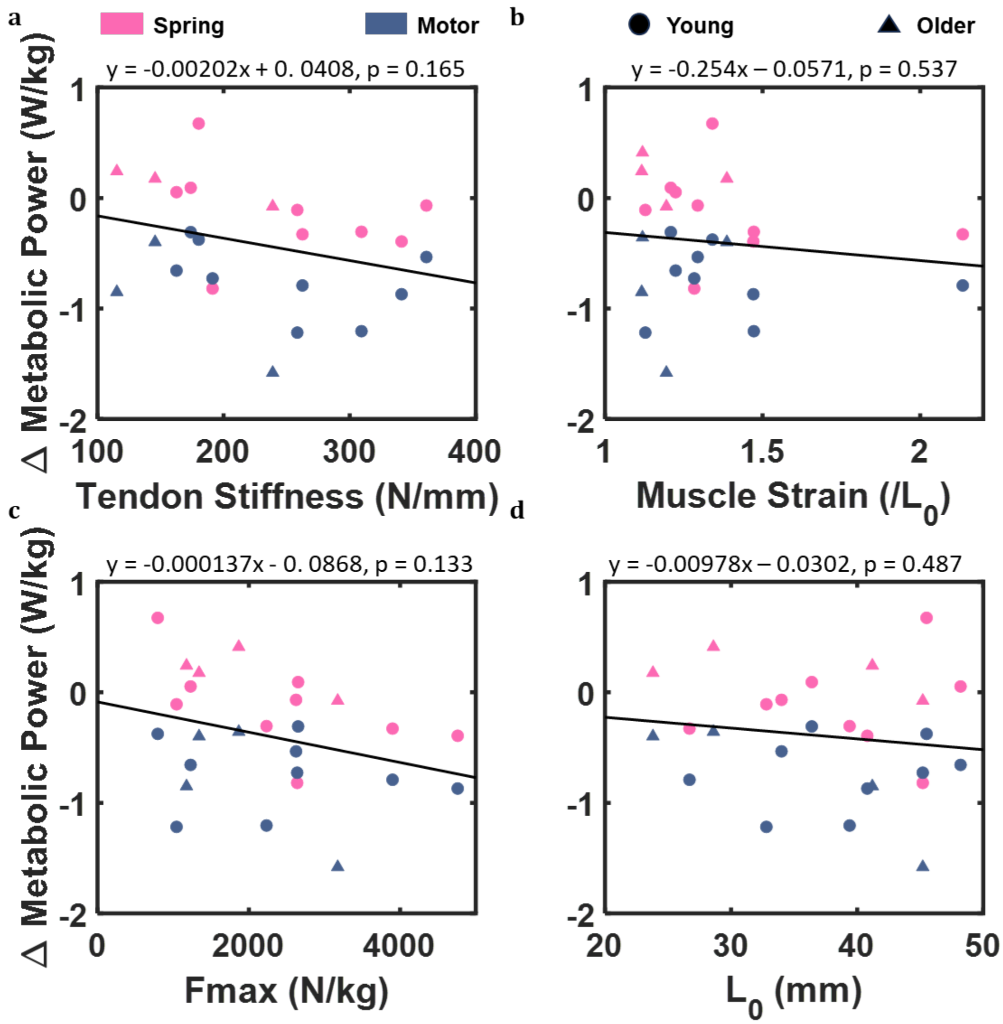

Figure 18 - Metabolic Power and Muscle-Tendon Properties: The relationship between the change in net metabolic power from the exoskeleton without assistance and a) Achilles tendon stiffness, b) passive muscle strain from zero to 20% of maximal force, c) maximal force, and d) optimal length at maximal activation for young (circle) and older (triangle) adults. The line is the linear regression fit to the data and the equation is above each corresponding plot. No correlation was found between the metabolic response to the exoskeleton and the muscle-tendon properties ($p > 0.05$).

power and tendon stiffness (Figure 18a). We also compared metabolic change to muscle strain, maximal force, and optimal length to see if exoskeleton performance was more related to muscle outcomes and again saw no significance (Figure 18). To our knowledge, no one has tried to link physiological properties to performance in exoskeletons. Structural properties of the ankle had no effect on exoskeleton performance potentially because in our previous work, we found the exoskeletons maintained ankle contribution while offloading the hip. Had we studied the hip musculature, we may have seen our hypothesized mechanism.

There are several limitations to this work. We did not optimize spring or motor assistance for the participants. Therefore, we did not tune the exoskeleton assistance to the participant's structure explaining the weak dependence of metabolic cost on structural properties. Without optimization of the springlike assistance, a reduction in metabolic cost was not found and further explaining no differences at the muscle level. Optimizing the motor, we may have seen larger reductions in metabolic cost and tuning towards individuals' structural properties emphasizing the changes at the muscle level. We also used aging to ensure a range of tendon stiffnesses. However, there is more than tendon stiffness that changes with age that we were unable to control in our study. Future studies that optimize exoskeleton assistance may better tweeze out how different types of torque profiles interact with the underlying physiology. Running the study with different populations that vary different muscle-tendon properties would further elucidate whether physiology impacts exoskeleton performance. Lastly, studies focusing on physiology at the hip may better predict exoskeleton performance since most of the changes kinetically were at the hip.

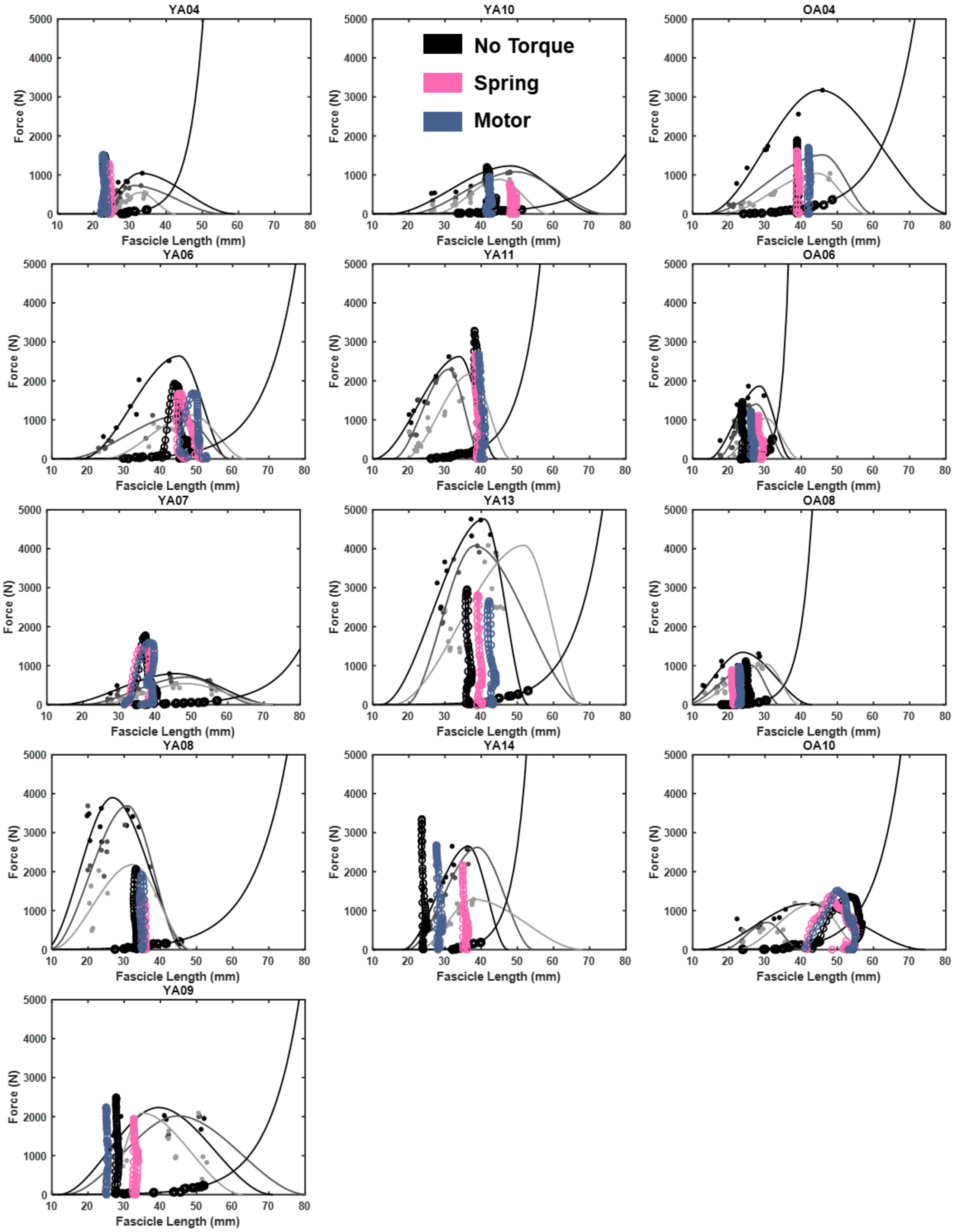

Figure 19 - Individual Absolute Force Length Curves: Absolute participant specific, activation dependent force-length curves with workloops for walking in exoskeletons without assistance (black), springlike assistance (pink), and motorlike assistance (blue).

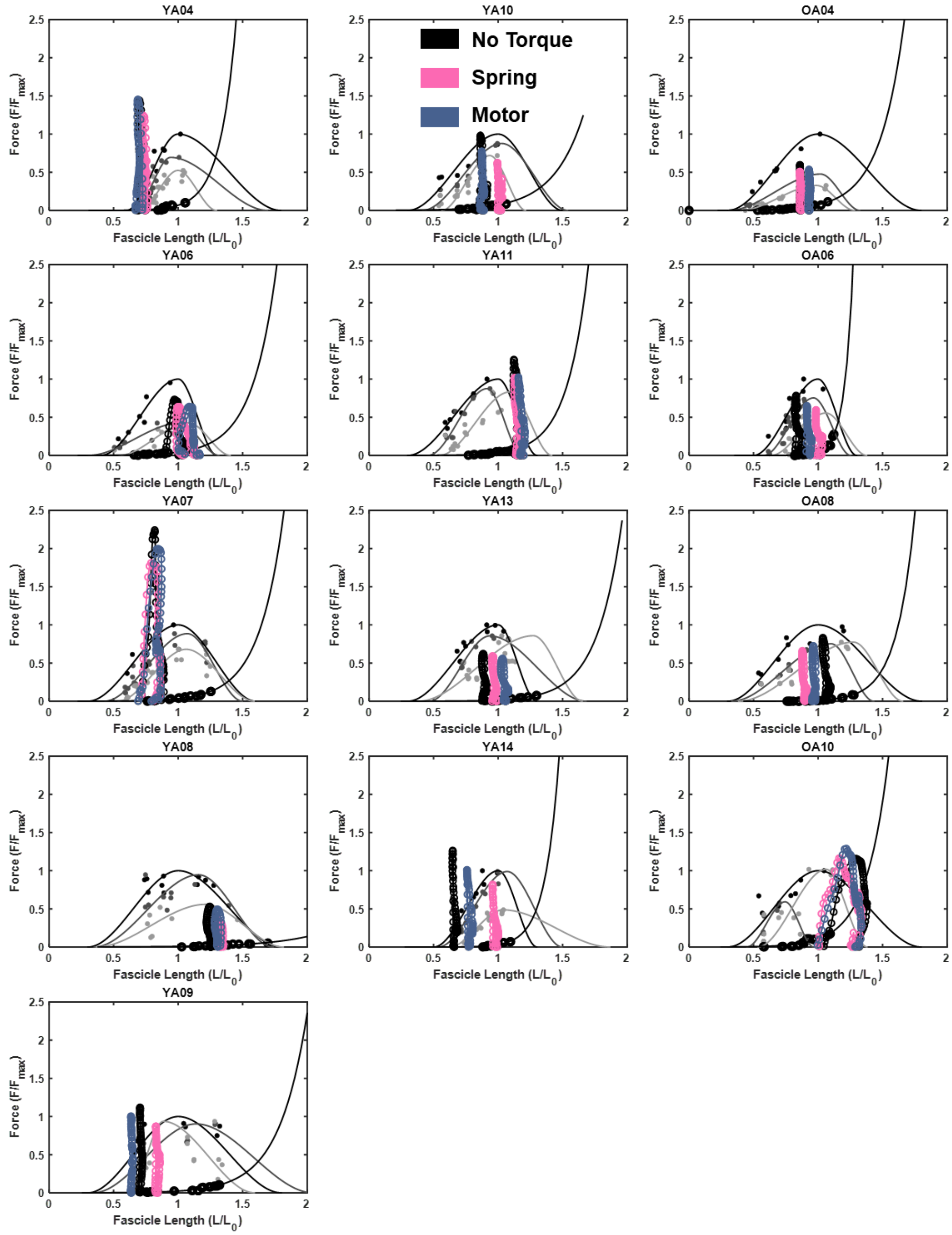

Figure 20 - Individual Normalized Force Length Curves: Normalized participant specific, activation dependent force-length curves with work loops for walking in exoskeletons without assistance (black), springlike assistance (pink), and motorlike assistance (blue).

CHAPTER 5. CONCLUSION

In this work we investigated the muscle and tendon changes with age, compared ankle exoskeleton assistance for young and older adults, and determined if exoskeleton assistance is influenced by calf muscle and tendon properties. In the first chapter we found both tendon stiffness and functional tendon stiffness were lower with age in agreement with literature. There is some debate where older adults operate on their tendon FL curve and whether less stiff tendon measures are just a function of operating at lower forces. The functional tendon stiffness shows that older adults operate with less stiff tendons at matched forces. While the muscle level properties were not different with age, we did have a fit group of older adults that don't represent an average older adult. This was also the first-time submaximal force-length curves were shown in older adults.

In chapter 2 we compared springlike and motorlike assistance in ankle exoskeletons and their ability to reduce metabolic cost. The comparison was two-fold, first we were able to see what type of assistance works better for older adults and secondly, the assistance profiles were previously tuned for young adults, allowing us to see if the assistance needed to be retuned for older adults or used across age groups. We found the motorlike assistance outperformed the springlike in both groups. This means focusing on motorlike assistance or assistance with late impulsive torque, should be used for older adults. We have not given up on passive devices, but a pseudo-passive device may be better than purely passive to achieve a more impulsive torque.

We also found that older adults were able to benefit from motorlike assistance optimized to young adults, meaning a one-size fits all approach worked in this case. The interesting component is that the motorlike assistance reduced metabolic cost not by only offloading the ankle, but rather maintaining or slightly reducing the ankle but primarily offloading the hip. While this has been seen in other studies, most ankle exoskeletons that reduce metabolic cost reduce the ankle contribution. This was also seen at the muscle level. Plantarflexion assistance during walking can assist in three main locations: 1) plantarflexion in late stance, 2) hip flexion in late stance, and 3) hip extension in early stance for the contralateral limb. The muscle activity was reduced in the ankle plantar flexors (GAS, SOL) in late stance, the hip flexors (RF) in late stance, and the hip extensors (BF, GMd, GMx) in early stance. This work shows how assistance can be redirected to assist other joints.

In chapter 3 we determined if calf muscle and tendon parameters influenced exoskeleton assistance performance. We found no correlation between metabolic power in the exoskeleton and muscle or tendon properties. There are a few ways this could be interpreted. 1) ankle muscle and tendon properties do not influence exoskeleton performance. 2) because the exoskeleton assistance wasn't tuned to the individuals' muscle and tendon properties, we did not see the interaction. 3a) the spring did not show a reduction in metabolic cost, so it is unlikely to see effects in our measurements. 3b) from chapter 2 we found the motorlike assistance primarily offloads the hip while maintaining ankle moment and power. With the ankle response being maintained, it is unlikely to see changes in the calf muscle and tendon.

We also found the reductions in metabolic cost with the motorlike assistance also came with a reduction in soleus fascicle force and activation while maintaining fascicle power. This means that the exoskeleton allows the participant to create the same amount of power while using less muscle activation justifying the lower metabolic cost. This is the first study to measure fascicle level changes while in an active exoskeleton.

With some answers, brings more questions. Future studies should see if tuning exoskeletons to older adults reveals greater reductions or shows new links between exoskeleton performance and physiological measures. It would also be interesting to see if other muscle or tendon measures would tell a different story or if studying the hip would show a link between physiology and exoskeleton performance. Since we saw improvements at the hip and ankle it would be interesting to see if a hip exoskeleton would be more beneficial. It would also be interesting to see how a pseudo-passive device could perform a more impulsive torque and if it could combine the benefits of both passive and active devices.

REFERENCES

[1] E. Rudnicka, P. Napierała, A. Podfigurna, B. Męczekalski, R. Smolarczyk, and M. Grymowicz, “The World Health Organization (WHO) approach to healthy ageing,” *Maturitas*, vol. 139, pp. 6–11, Sep. 2020, doi: 10.1016/j.maturitas.2020.05.018.

[2] C. A. Okoro, N. D. Hollis, A. C. Cyrus, and S. Griffin-Blake, “Morbidity and Mortality Weekly Report Prevalence of Disabilities and Health Care Access by Disability Status and Type Among Adults-United States, 2016,” *Morbidity and Mortality Weekly Report*, 2018, [Online]. Available: https://aspe.hhs.gov/datacncl/standards/aca/4302/index.pdf.

[3] “Older Americans 2020 Key Indicators of Well-Being,” 2020.

[4] J. D. Ortega and C. T. Farley, “Individual limb work does not explain the greater metabolic cost of walking in elderly adults,” *J Appl Physiol*, vol. 102, no. 6, pp. 2266–2273, 2007, doi: 10.1152/japplphysiol.00583.2006.

[5] J. E. Himann, D. A. Cunningham, P. A. Rechnitzer, and D. H. Paterson, “Age-related changes in speed of walking,” *Med Sci Sports Exerc*, vol. 20, no. 2, 1988.

[6] S. Morrison *et al.*, “Walking-Induced Fatigue Leads to Increased Falls Risk in Older Adults,” *J Am Med Dir Assoc*, vol. 17, no. 5, pp. 402–409, May 2016, doi: 10.1016/j.jamda.2015.12.013.

[7] L. D. S. S. C. B. de Oliveira, E. C. Souza, R. A. S. Rodrigues, C. A. Fett, and A. B. Piva, “The effects of physical activity on anxiety, depression, and quality of life in elderly people living in the community,” *Trends Psychiatry Psychother*, vol. 41, no. 1, pp. 36–42, Jan. 2019, doi: 10.1590/2237-6089-2017-0129.

[8] J. R. Franz and R. Kram, “Advanced age and the mechanics of uphill walking: A joint-level, inverse dynamic analysis,” *Gait Posture*, vol. 39, no. 1, pp. 135–140, 2014, doi: 10.1016/j.gaitpost.2013.06.012.

[9] T. Delabastita, E. Hollville, A. Catteau, P. Cortvriendt, F. De Groote, and B. Vanwanseele, “Distal-to-proximal joint mechanics redistribution is a main

contributor to reduced walking economy in older adults," *Scand J Med Sci Sports*, 2021, doi: 10.1111/sms.13929.

[10] R. E. Pimentel, J. N. Feldman, M. D. Lewek, and J. R. Franz, "Quantifying Cause-Effect Relations Between Walking Speed, Propulsive Force, and Metabolic Cost", doi: 10.1101/2021.10.18.21265129.

[11] G. S. Sawicki, C. L. Lewis, and D. P. Ferris, "It pays to have springs in your step," *Exerc Sport Sci Rev*, vol. 37, no. 3, pp. 1–19, 2013, doi: 10.1097/JES.0b013e31819c2df6.It.

[12] R. L. Krupenevich, O. N. Beck, G. S. Sawicki, and J. R. Franz, "Reduced Achilles Tendon Stiffness Disrupts Calf Muscle Neuromechanics in Elderly Gait," *Gerontology*, 2021, doi: 10.1159/000516910.

[13] K. J. Gries and S. Trappe, "The Aging Athlete: Paradigm of Healthy Aging," *Int J Sports Med*, Feb. 2022, doi: 10.1055/a-1761-8481.

[14] G. L. Onambele, M. V. Narici, and C. N. Maganaris, "Calf muscle-tendon properties and postural balance in old age," *J Appl Physiol*, vol. 100, no. 6, pp. 2048–2056, 2006, doi: 10.1152/japplphysiol.01442.2005.

[15] I. Lindemann, B. K. Coombes, K. Tucker, F. Hug, and J. M. Dick, "Age-related differences in gastrocnemii muscles and Achilles tendon mechanical properties in vivo," *J Biomech*, vol. 112, p. 110067, 2020, doi: 10.1016/j.jbiomech.2020.110067.

[16] P. Lopez *et al.*, "Benefits of resistance training in physically frail elderly: a systematic review," *Aging Clin Exp Res*, vol. 30, no. 8, pp. 889–899, Aug. 2018, doi: 10.1007/s40520-017-0863-z.

[17] S. W. Han, D. Y. Lee, D. S. Choi, B. Han, J. S. Kim, and H. D. Lee, "Asynchronous alterations of muscle force and tendon stiffness following 8 weeks of resistance exercise with whole-body vibration in older women," *J Aging Phys Act*, vol. 25, no. 2, pp. 287–294, Apr. 2017, doi: 10.1123/japa.2016-0149.

[18] J. Grgic, A. Garofolini, J. Orazem, F. Sabol, B. J. Schoenfeld, and Z. Pedisic, "Effects of Resistance Training on Muscle Size and Strength in Very Elderly Adults: A Systematic Review and Meta-Analysis of Randomized Controlled Trials," *Sports*

Medicine, vol. 50, no. 11. Springer Science and Business Media Deutschland GmbH, pp. 1983–1999, Nov. 01, 2020. doi: 10.1007/s40279-020-01331-7.

[19] G. S. Sawicki, O. N. Beck, I. Kang, and A. J. Young, "The exoskeleton expansion: Improving walking and running economy," *J Neuroeng Rehabil*, vol. 17, no. 1, pp. 1–9, 2020, doi: 10.1186/s12984-020-00663-9.

[20] P. W. Franks, G. M. Bryan, R. M. Martin, R. Reyes, and S. H. Collins, "Comparing optimized exoskeleton assistance of the hip , knee , and ankle in single and multi-joint configurations," *BioRxiv*, 2021.

[21] P. Slade, M. J. Kochenderfer, S. L. Delp, and S. H. Collins, "Personalizing exoskeleton assistance while walking in the real world," *Nature*, vol. 610, no. 7931, pp. 277–282, Oct. 2022, doi: 10.1038/s41586-022-05191-1.

[22] S. H. Collins, M. Bruce Wiggin, and G. S. Sawicki, "Reducing the energy cost of human walking using an unpowered exoskeleton," *Nature*, vol. 522, no. 7555, pp. 212–215, 2015, doi: 10.1038/nature14288.

[23] R. W. Nuckols and G. S. Sawicki, "Impact of elastic ankle exoskeleton stiffness on neuromechanics and energetics of human walking across multiple speeds," *J Neuroeng Rehabil*, pp. 1–50, 2020, doi: 10.21203/rs.2.20510/v1.

[24] J. Zhang *et al.*, "Human-in-the-Loop Optimization of Exoskeleton Assistance During Walking," *Science Report*, vol. 356, pp. 1280–1284, 2017, doi: 10.1016/j.gaitpost.2011.08.025.The.

[25] D. R. Louie and J. J. Eng, "Powered robotic exoskeletons in post-stroke rehabilitation of gait: A scoping review," *Journal of NeuroEngineering and Rehabilitation*, vol. 13, no. 1. BioMed Central Ltd., Jun. 08, 2016. doi: 10.1186/s12984-016-0162-5.

[26] E. Guanziroli, M. Cazzaniga, L. Colombo, S. Basilico, G. Legnani, and F. Molteni, "Assistive powered exoskeleton for complete spinal cord injury : correlations between walking ability and exoskeleton control exoskeleton control Corresponding author :," *Eur J Phys Rehabil Med*, 2018, doi: 10.23736/S1973-9087.18.05308-X.

[27] B. C. Conner, G. Orekhov, and Z. Lerner, "Ankle exoskeleton assistance increases six-minute walk test performance in cerebral palsy," *IEEE Open J Eng Med Biol*, pp. 1–1, 2021, doi: 10.1109/OJEMB.2021.3135826.

[28] F. A. Panizzolo, C. Bolgiani, L. Di Liddo, E. Annese, and G. Marcolin, "Reducing the energy cost of walking in older adults using a passive hip flexion device," *J Neuroeng Rehabil*, vol. 16, no. 1, Oct. 2019, doi: 10.1186/s12984-019-0599-4.

[29] H. J. Lee *et al.*, "A Wearable Hip Assist Robot Can Improve Gait Function and Cardiopulmonary Metabolic Efficiency in Elderly Adults," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 25, no. 9, pp. 1549–1557, Sep. 2017, doi: 10.1109/TNSRE.2017.2664801.

[30] S. Galle, W. Derave, F. Bossuyt, P. Calders, P. Malcolm, and D. De Clercq, "Exoskeleton plantar flexion assistance for elderly," *Gait Posture*, vol. 52, pp. 183–188, 2017.

[31] D. A. Winter, "Kinematic and kinetic patterns in human gait: Variability and compensating effects," *Hum Mov Sci*, vol. 3, no. 1–2, pp. 51–76, 1984, doi: 10.1016/0167-9457(84)90005-8.

[32] O. N. Beck, L. H. Trejo, J. N. Schroeder, J. R. Franz, and G. S. Sawicki, "Shorter muscle fascicle operating lengths increase the metabolic cost of cyclic force production," *J Appl Physiol*, vol. 133, no. 3, pp. 524–533, Sep. 2022, doi: 10.1152/japplphysiol.00720.2021.

[33] T. J. Roberts and E. Azizi, "Flexible mechanisms: the diverse roles of biological springs in vertebrate movement," *Journal of Experimental Biology*, vol. 214, no. 3, pp. 353–361, 2011, doi: 10.1242/jeb.038588.

[34] J. Rubenson, N. J. Pires, H. O. Loi, G. J. Pinniger, and D. G. Shannon, "On the ascent: the soleus operating length is conserved to the ascending limb of the force-length curve across gait mechanics in humans," *Journal of Experimental Biology*, vol. 215, no. 20, pp. 3539–3551, 2012, doi: 10.1242/jeb.070466.

[35] N. C. Holt and E. Azizi, "What drives activation-dependent shifts in the force-length curve ?," *Biol Lett*, vol. 10, 2014, doi: 10.1098/rsbl.2014.0651.

[36] H. D. B. Fontana and W. Herzog, "Vastus lateralis maximum force - generating potential occurs at optimal fascicle length regardless of activation level," *Eur J Appl Physiol*, vol. 116, no. 6, pp. 1267–1277, 2016, doi: 10.1007/s00421-016-3381-3.

[37] A. Monte, P. Tecchio, F. Nardello, B. Bachero-Mena, L. P. Ardigò, and P. Zamparo, "The interplay between gastrocnemius medialis force–length and force–velocity potentials, cumulative EMG activity and energy cost at speeds above and below the walk to run transition speed," *Exp Physiol*, vol. 108, no. 1, pp. 90–102, Jan. 2023, doi: 10.1113/EP090657.

[38] F. A. Panizzolo, D. J. Green, D. G. Lloyd, A. J. Maiorana, and J. Rubenson, "Soleus fascicle length changes are conserved between young and old adults at their preferred walking speed," *Gait Posture*, vol. 38, no. 4, pp. 764–769, Sep. 2013, doi: 10.1016/j.gaitpost.2013.03.021.

[39] C. I. Morse, J. M. Thom, K. M. Birch, and M. V Narici, "Changes in triceps surae muscle architecture with sarcopenia," *Acta Physiol Scand*, 2005.

[40] N. C. Holt, N. Danos, T. J. Roberts, and E. Azizi, "Stuck in gear: age-related loss of variable gearing in skeletal muscle," *J Exp Biol*, vol. 219, no. 7, pp. 998–1003, 2016, doi: 10.1242/jeb.133009.

[41] J. R. Franz and D. G. Thelen, "Depth-dependent variations in Achilles tendon deformations with age are associated with reduced plantarflexor performance during walking," *J Appl Physiol*, vol. 119, no. 3, pp. 242–249, 2015, doi: 10.1152/japplphysiol.00114.2015.

[42] M. CN, "Force-length characteristics of in vivo human skeletal muscle," *Acta Physiol Scand*, vol. 172, no. 4, p. 279, 2001, [Online]. Available: http://search.ebscohost.com.proxy-ub.rug.nl/login.aspx?direct=true&db=amed&AN=0029027&site=ehost-live&scope=site

[43] A. L. Hessel, B. J. Raiteri, M. J. Marsh, and D. Hahn, "Rightward shift of optimal fascicle length with decreasing voluntary activity level in the soleus and lateral gastrocnemius muscles," *Journal of Experimental Biology*, vol. 224, Jan. 2021, doi: 10.1242/jeb.235614.

[44] K. Manal, J. D. Cowder, and T. S. Buchanan, “A Hybrid Method for Computing Achilles Tendon Moment Arm Using Ultrasound and Motion Analysis,” *J Appl Biomech*, vol. 26, no. 2, pp. 224–228, 2010, doi: 10.1038/jid.2014.371.

[45] D. A. Winter, *Biomechanics and Motor Control of Human Movement Fourth Edition*, 4th ed., vol. 2nd. 2009. doi: 10.1002/9780470549148.

[46] D. J. Farris and G. A. Lichtwark, “UltraTrack: Software for semi-automated tracking of muscle fascicles in sequences of B-mode ultrasound images.,” *Comput Methods Programs Biomed*, vol. 128, pp. 111–8, 2016, doi: 10.1016/j.cmpb.2016.02.016.

[47] S. K. Gollapudi and D. C. Lin, “Experimental determination of sarcomere force-length relationship in type-I human skeletal muscle fibers,” *J Biomech*, vol. 42, no. 13, pp. 2011–2016, Sep. 2009, doi: 10.1016/j.jbiomech.2009.06.013.

[48] V. Baltzopoulos, J. G. Williams, and D. A. Brodie, “Sources of Error in Isokinetic Dynamometry: Effects of Visual Feedback on Maximum Torque Measurements,” 1991. [Online]. Available: www.jospt.org

[49] S. Bohm, F. Mersmann, M. Tettke, M. Kraft, and A. Arampatzis, “Human Achilles tendon plasticity in response to cyclic strain: Effect of rate and duration,” *Journal of Experimental Biology*, vol. 217, no. 22, pp. 4010–4017, Nov. 2014, doi: 10.1242/jeb.112268.

[50] A. Arampatzis, A. Peper, S. Bierbaum, and K. Albracht, “Plasticity of human Achilles tendon mechanical and morphological properties in response to cyclic strain,” *J Biomech*, vol. 43, no. 16, pp. 3073–3079, Dec. 2010, doi: 10.1016/j.jbiomech.2010.08.014.

[51] S. L. Delp *et al.*, “OpenSim : Open-Source Software to Create and Analyze Dynamic Simulations of Movement,” *IEEE Trans Biomed Eng*, vol. 54, no. 11, pp. 1940–1950, 2007.

[52] T. Fukunaga *et al.*, “Physiological cross-sectional area of human leg muscles based on magnetic resonance imaging,” *Journal of Orthopaedic Research*, vol. 10, no. 6, pp. 926–934, 1992, doi: 10.1002/jor.1100100623.

[53] N. C. Holt and E. Azizi, “What drives activation-dependent shifts in the force-length curve ?,” *Biol Lett*, vol. 10, 2014, doi: 10.1098/rsbl.2014.0651.

[54] N. C. Holt and E. Azizi, “The effect of activation level on muscle function during locomotion: Are optimal lengths and velocities always used?,” *Proceedings of the Royal Society B*, vol. 283, no. 1823, 2016, doi: 10.1098/rspb.2015.2832.

[55] R. L. Lieber and S. R. Ward, “Skeletal muscle design to meet functional demands,” *Philosophical Transactions of the Royal Society B: Biological Sciences*, vol. 366, no. 1570, pp. 1466–1476, 2011, doi: 10.1098/rstb.2010.0316.

[56] J. R. Franz, “The Age-Associated Reduction in Propulsive Power Generation in Walking,” *Exerc Sport Sci Rev*, vol. 44, no. 4, pp. 129–136, 2016, doi: 10.1249/JES.0000000000000086.

[57] P. DeVita and T. Hortobagyi, “Age causes a redistribution of joint torques and powers during gait.,” *J Appl Physiol (1985)*, vol. 88, no. 5, pp. 1804–1811, 2000, doi: 10.1152/jappl.2000.88.5.1804.

[58] K. Kubo, Y. Ishida, T. Komuro, N. Tsunoda, H. Kanehisa, and T. Fukunaga, “Age-Related Differences in the Force Generation Capabilities and Tendon Extensibilities of Knee Extensors and Plantar Flexors in Men,” *J Gerontol A Biol Sci Med Sci*, vol. 62, no. 11, pp. 1252–1258, Nov. 2007, doi: 10.1093/gerona/62.11.1252.

[59] L. Hou *et al.*, “Effect of Protein Supplementation Combined With Resistance Training on Muscle Mass, Strength and Function in the Elderly: A Systematic Review and Meta-Analysis,” *Journal of Nutrition, Health and Aging*, vol. 23, no. 5, pp. 451–458, May 2019, doi: 10.1007/s12603-019-1181-2.

[60] K. Kubo, M. Morimoto, T. Komuro, N. Tsunoda, H. Kanehisa, and T. Fukunaga, “Age-related differences in the properties of the plantar flexor muscles and tendons,” *Med Sci Sports Exerc*, vol. 39, no. 3, pp. 541–547, Mar. 2007, doi: 10.1249/01.mss.0000247006.24965.74.

[61] S. Bohm, F. Mersmann, and A. Arampatzis, “Human tendon adaptation in response to mechanical loading: a systematic review and meta-analysis of exercise intervention studies on healthy adults,” *Sports Med Open*, vol. 1, no. 1, 2015, doi: 10.1186/s40798-015-0009-9.

[62] M. Grimmer, R. Riener, C. J. Walsh, and A. Seyfarth, "Mobility related physical and functional losses due to aging and disease - A motivation for lower limb exoskeletons," *J Neuroeng Rehabil*, vol. 16, no. 1, Jan. 2019, doi: 10.1186/s12984-018-0458-8.

[63] P. W. Franks, G. M. Bryan, R. M. Martin, R. Reyes, A. C. Lakmazaheri, and S. H. Collins, "Comparing optimized exoskeleton assistance of the hip, knee, and ankle in single and multi-joint configurations," *Wearable Technologies*, vol. 2, p. e16, Nov. 2021, doi: 10.1017/wtc.2021.14.

[64] Y. Fang, K. Harshe, J. R. Franz, and Z. F. Lerner, "Feasibility evaluation of a dual-mode ankle exoskeleton to assist and restore community ambulation in older adults," *Wearable Technologies*, vol. 3, Jul. 2022, doi: 10.1017/wtc.2022.12.

[65] K. L. Poggensee and S. H. Collins, "How adaptation, training, and customization contribute to benefits from exoskeleton assistance," *Sci Robot*, vol. 6, no. 58, Sep. 2021, doi: 10.1126/scirobotics.abf1078.

[66] J. M. Brockway, "Derivation of formulae used to calculate energy expenditure in man," *Hum Nutr Clin Nutr*, vol. 41, no. 6, pp. 463–71, 1987, doi: 10.1016/j.pcad.2015.11.006.

[67] R. W. Jackson and S. H. Collins, "An experimental comparison of the relative benefits of work and torque assistance in ankle exoskeletons," *J Appl Physiol*, vol. 119, no. 5, pp. 541–557, 2015, doi: 10.1152/japplphysiol.01133.2014.

[68] S. Galle, P. Malcolm, S. H. Collins, and D. De Clercq, "Reducing the metabolic cost of walking with an ankle exoskeleton: interaction between actuation timing and power," *J Neuroeng Rehabil*, vol. 14, no. 1, pp. 1–16, 2017, doi: 10.1186/s12984-017-0235-0.

[69] B. D. Robertson and G. S. Sawicki, "Exploiting elasticity: Modeling the influence of neural control on mechanics and energetics of ankle muscle-tendons during human hopping," *J Theor Biol*, vol. 353, pp. 121–132, 2014, doi: 10.1016/j.jtbi.2014.03.010.

[70] B. D. Robertson, D. J. Farris, and G. S. Sawicki, "More is not always better: Modeling the effects of elastic exoskeleton compliance on underlying ankle muscle-

tendon dynamics," *Bioinspir Biomim*, vol. 9, no. 4, p. 46018, 2014, doi: 10.1088/1748-3182/9/4/046018.

[71] G. S. Sawicki, B. D. Robertson, E. Azizi, and T. J. Roberts, "Timing matters: Tuning the mechanics of a muscle-tendon unit by adjusting stimulation phase during cyclic contractions," *Journal of Experimental Biology*, vol. 218, no. 19, pp. 3150–3159, Oct. 2015, doi: 10.1242/jeb.121673.

[72] L. M. Mooney and H. M. Herr, "Biomechanical walking mechanisms underlying the metabolic reduction caused by an autonomous exoskeleton," *J Neuroeng Rehabil*, vol. 13, no. 1, pp. 1–12, 2016, doi: 10.1186/s12984-016-0111-3.

[73] J. R. Koller, D. A. Jacobs, D. P. Ferris, and C. D. Remy, "Learning to walk with an adaptive gain proportional myoelectric controller for a robotic ankle exoskeleton," *J Neuroeng Rehabil*, vol. 12, no. 1, pp. 1–14, 2015, doi: 10.1186/s12984-015-0086-5.

[74] M. I. V. Orselli, J. R. Franz, and D. G. Thelen, "The effects of Achilles tendon compliance on triceps surae mechanics and energetics in walking," *J Biomech*, vol. 60, pp. 227–231, 2017, doi: 10.1016/j.jbiomech.2017.06.022.

[75] C. Couppé *et al.*, "Human Achilles tendon glycation and function in diabetes," vol. 9, no. 23, pp. 130–137, 2021, doi: 10.1152/japplphysiol.00547.2015.

[76] M. Petrovic *et al.*, "Altered Achilles tendon function during walking in people with diabetic neuropathy: Implications for metabolic energy saving," *J Appl Physiol*, vol. 124, no. 5, pp. 1333–1340, 2018, doi: 10.1152/japplphysiol.00290.2017.

[77] L. Barber, R. Barrett, and G. Lichtwark, "Medial gastrocnemius muscle fascicle active torque-length and Achilles tendon properties in young adults with spastic cerebral palsy," *J Biomech*, vol. 45, no. 15, pp. 2526–2530, Oct. 2012, doi: 10.1016/j.jbiomech.2012.07.018.

[78] D. J. Farris and G. S. Sawicki, "Linking the mechanics and energetics of hopping with elastic ankle exoskeletons," *J Appl Physiol*, vol. 113, no. 12, pp. 1862–1872, 2012, doi: 10.1152/japplphysiol.00802.2012.

[79] R. W. Nuckols, T. J. M. Dick, O. N. Beck, and G. S. Sawicki, “Ultrasound imaging links soleus muscle neuromechanics and energetics during human walking with elastic ankle exoskeletons,” *Nature Scientific Reports*, vol. 10, no. 3604, pp. 1–15, 2020, doi: 10.1038/s41598-020-60360-4.

[80] R. W. Jackson, C. L. Dembia, S. L. Delp, and S. H. Collins, “Muscle–tendon mechanics explain unexpected effects of exoskeleton assistance on metabolic rate during walking,” *J Exp Biol*, vol. 220, no. 11, pp. 2082–2095, 2017, doi: 10.1242/jeb.150011.