THE RELATIONSHIPS BETWEEN PHYSICAL CAPACITY AND BIOMECHANICAL PLASTICITY WITH AGE DURING LEVEL AND INCLINE WALKING

By

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April, 2017

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Old compared to young adults exhibit a distal-to-proximal redistribution of joint contributions to support phase mechanics during level and incline walking. Although this biomechanical plasticity is now well established in healthy old adults, less is known about how biomechanical plasticity varies across the physical capacity spectrum in this population. For example, it is unclear whether high capacity old adults (i.e. individuals with relatively high walking performance) retain a more youthful gait strategy (i.e. low magnitudes of biomechanical plasticity) or adopt larger magnitudes of biomechanical plasticity in order to walk well. The purpose of my thesis was to examine and quantify the relationships between physical capacity and biomechanical plasticity in old adults during level and incline walking. We hypothesized that, as physical capacity declines, biomechanical plasticity would increase in magnitude. We also hypothesized that the magnitude of change in biomechanical plasticity per unit change in physical capacity would be greater during the more challenging task of incline compared to level walking. To test these hypotheses, we performed gait analyses on 10 young and 32 old adults as they walked over level and inclined (+10°) surfaces at self-selected and controlled speeds. We used Short-Form Health Survey Physical Component (SF-36 PC) scores and 20-meter self-selected speeds as measures of physical capacity. To quantify biomechanical plasticity, we created ratios of hip extensor to ankle plantarflexor peak torques, angular impulses, peak positive powers, and positive work. Compared to young adults, old adults exhibited larger biomechanical

plasticity ratios during all four walking conditions – confirming the existence of age-associated biomechanical plasticity in the old adults included in this study. Contrary to our hypothesis, correlation analyses revealed positive relationships between physical capacity and biomechanical plasticity during level and incline walking at self-selected but not controlled speeds. Positive relationships between in-trial self-selected speeds during level walking suggest that increased magnitudes of biomechanical plasticity might positively influence walking performance. Contrary to our second hypothesis, incline walking did not increase magnitude of biomechanical plasticity change per unit change in physical capacity. Our results suggest that age-associated biomechanical plasticity represents a beneficial gait adaptation that might afford functional benefits such as increased walking speed. Results from our cross-sectional design may provide the framework for a longitudinal intervention study aimed at increasing biomechanical plasticity and thereby walking performance in old adults. Increased walking performance in this population has the potential to decrease adverse outcomes such as falls, hospitalizations, and even mortality, leading to an overall increased quality of life.

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A Thesis

Presented to the Faculty of the Department of Kinesiology

East Carolina University

In Partial Fulfillment of the Requirements for the Degree

Masters of Science in Kinesiology

Biomechanics Option

By

Daniel James Kuhman

April, 2017

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Chapter I: Introduction

Introduction

Healthy and pathological aging are associated with decreased skeletal muscle mass[49], strength[49,86], and power[5,32]. These age-associated reductions to muscular function can have a profound impact on the ability of old adults to perform daily activities such as walking and can result in a reduced quality of life. Indeed, reduced walking capacity in old adults, including shorter step length[115] and slower walking speed[11], is predicative of falls, disability, hospitalization, and even mortality[1,82,105]. Because of the consequences of decreased muscle quality and subsequent decline in walking performance, the underlying biomechanical components of walking gait in old adults have been examined.

Compared to young adults, healthy old adults exhibit increased hip joint extensor torque and positive power during the early support phase, increased hip flexor torque in late support[59], and decreased ankle joint plantarflexor torque and positive power[24,59,92,102,115] during the late support phase of level walking. This distal-to-proximal redistribution of joint contributions was precisely quantified by DeVita & Hortobagyi (2000), who reported that, while walking at the same speed (1.5 ms^{-1}), hip and ankle contributions to total positive joint work were 44% and 51% respectively for old adults and 16% and 73% respectively for young adults[24]. Redistribution of joint contributions has been termed biomechanical plasticity and is now considered a fundamental principle that quantifies altered joint-level mechanics that occur with altered states of health. In addition to the distal-to-proximal redistribution pattern of joint contributions in old adults, biomechanical plasticity has been observed in ACL injured adults[25,26], morbidly obese adults both before and after weight loss[23,55], and individuals with multiple sclerosis[20]. Age-associated biomechanical plasticity has also been observed during incline ascent, where old adults exhibit increased hip and decreased ankle joint contributions during the support phase compared to young adults[37]. However, at a 9° incline, old adults do not increase ankle joint torque compared to level walking[37], suggesting a more pronounced biomechanical plasticity, possibly due to increased task difficulty. Decreased muscular function, particularly of the ankle plantarflexors[69], in old adults is likely a major component of this age-associated biomechanical plasticity during locomotion. Indeed, Hortobagyi, et al. (2016) reported increased hip and decreased ankle joint relative contributions to total positive joint work in weak (48.5% and 38.7%, respectively) compared to strong (41.3% and 46.6%, respectively) old adults[56]. More difficult tasks, such as ascending inclines, that place a greater load on the muscles of old individuals relative to their functional capacity, may cause biomechanical plasticity to become more pronounced.

The previously discussed gait adaptations pertain to *healthy* old adults. However, not all adults age in a healthy and robust manner. To determine the effects of physical capacity on gait adaptations, comparisons have been made between healthy (high-capacity) and *low-capacity* old adults. Compared to healthy old adults, low-capacity old adults walk with slower gait speeds[71], the importance of which was discussed previously. Joint-level biomechanical comparisons between high and low-capacity old adults show that the same age-associated biomechanical plasticity exists in both groups, however the magnitude of plasticity appears to be larger in low-capacity groups. More specifically, low-capacity old adults exhibit increased peak hip extensor power during early support[52,71,72] and peak flexor torque and power in late support[71], as well as decreased peak plantarflexor power and work during late support compared to healthy old adults[52,71,72]. Currently, no study has directly compared high and low-capacity old adults during incline walking. However, given that incline walking requires increased power generation from both the hip and ankle joints[37], and the apparent lack of power generation capacity in the ankle joint of low-capacity old adults, it is likely that the low-capacity group will rely even more on hip and even less on ankle joint contributions compared to healthy old adults during incline walking. Further decline of muscular function in low compared to high-capacity old adults is likely a major component of the increased magnitude of plasticity in the low-capacity groups. All of these results indicate that low-capacity old adults exhibit larger magnitudes of age-associated biomechanical plasticity compared to healthy old adults, perhaps due to further muscular decline. However, all of these studies separated healthy and low-capacity old adults into two discrete bins for statistical comparisons. Thus far, no study has examined age-associated biomechanical plasticity while treating physical capacity as a continuous, rather than discrete, variable.

Hypothesis

Based on the previous research, it is hypothesized that physical capacity and age-associated biomechanical plasticity are inversely related such that, as physical capacity declines, the magnitude of age-associated biomechanical plasticity increases. More specifically, we hypothesize that as physical capacity declines, old adults will rely more heavily on hip joint contributions and less on ankle joint contributions during the support phase of walking. It is also hypothesized that the magnitude of biomechanical plasticity becomes more pronounced during the more challenging task of incline walking. In other words, the increase in magnitude of biomechanical plasticity per unit change of physical capacity will be higher during incline compared to level walking.

Purpose

The purpose of this thesis is to examine the relationships between physical capacity and age-associated biomechanical plasticity during level and incline walking. The results will precisely and mathematically quantify the relationships between physical capacity and biomechanical plasticity during both level and incline walking.

Significance

Understanding the relationship between physical capacity and biomechanical plasticity in old adults will allow us to view gait adaptations along a biological spectrum rather than chronological timeline. Previous research suggests that biomechanical plasticity in old adults occurs, at least in part, due to declining muscular function. Although some amount of muscular decline is naturally associated with age, the amount of muscular decline per individual is more of a biological rather than chronological consequence. This idea is supported by the increased magnitude of biomechanical plasticity reported in low-capacity and weak compared to healthy and strong old adults in age-matched samples. In this regard, results from the current study will provide a novel and precise mathematical assessment of the amount of biomechanical plasticity associated with reduced physical capacity in old adults. The findings may also help guide the development and implementation of training programs aimed at old populations of differing physical capacities. Successful training programs may increase walking capacity and quality of life in old adults, as well as reduce the incidence of falls, disabilities, dependencies, hospitalizations, and mortalities in this population.

Delimitations

1. Young participants will include males and females aged 18-25 without a history of lower limb pain or injury, neuromuscular or musculoskeletal complications, other orthopedic problems, or history of any neurological limitations (stroke, Parkinson's Disease, etc.).

2. Old participants will include both males and females aged 70-85 years of varying physical capacities.

3. All participants will have a Body Mass Index of less than 30 kg/m^2 to control for possible obesity effects on gait.

4. This study is limited to the analysis of the support phase of the right lower limb during 5 trials of each of 4 conditions.

5. The analysis will focus only on biomechanical characteristics of level and incline walking gaits.

Operational definitions

1. Age-associated biomechanical plasticity in locomtoion: The distal-to-proximal shift in joint kinetics (torques, angular impulse, powers, work) in old adults. More specifically, the increased hip joint kinetics during early support and decreased ankle joint kinetics during late support of walking in old compared to young adults.

2. Physical capacity: A measure of one's ability to perform activities commonly encountered in daily life. We will quantify physical capacity using SF-36 Physical Component (PC) scores and self-selected walking speed over a 20-meter level walkway.

3. Young adults: Males and females aged 18-25 years.

4. Old adults: Males and females aged 70-85 years.

5. Proximal Muscles: Muscles acting at the hip joint.

6. Distal Muscles: Muscles acting at the knee and ankle joints.

7. Support torque: the sum of the sagittal plane hip, knee, and ankle joint torques during the support phase of walking.

8. Total power – the sum of the hip, knee, and ankle joint powers during the support phase of level walking.

Chapter II: A review of the literature

Introduction

The purpose of the current study is to examine the relationships between physical capacity and age-associated biomechanical plasticity during level and incline walking. The following literature review will discuss: healthy gait biomechanics – level & incline, declining muscular function with age, biomechanical plasticity in old adults, biomechanical plasticity in low-capacity old adults, defining and measuring physical capacity, walking gait alterations with decreasing physical capacity, the hypothesis and purpose, and a summary.

Healthy gait biomechanics – level & incline

Bipedal walking is an activity performed by most people every day. Although it is typically viewed as a relatively simple activity, it requires complex and coordinated movements that simultaneously aid in forward movement of the center of mass and full-body stability. To better understand healthy walking gait, its biomechanical components have been extensively analyzed.

Typical human walking can be broadly divided into a swing phase, during which the foot is not in contact with the ground, and a support phase, during which the foot is in contact with the ground, for each individual leg. The swing phase, also termed limb advancement, can be further divided into initial swing, mid-swing, and terminal swing phases[27]. The support phase can be further divided into weight acceptance (initial ground contact and the loading response) and limb support (mid-support, terminal support, and pre-swing) phases[27]. Healthy individuals spend approximately 40% of the gait cycle in the swing phase and the remaining 60% in the support phase. At typical walking speeds, a double support phase, in which both feet are in contact with the ground simultaneously, is present. Each of these walking gait phases are comprised of distinct kinematic and kinetic patterns, however the remainder of this report will focus primarily on support phase biomechanics.

Kinematics refer to motions without reference to the forces that cause said motions. These variables include, among other things, angular positions and angular velocities at skeletal joints. Joint angular positions in young, healthy adults walking over level ground exhibit consistently similar hip, knee, and ankle sagittal plane movement patterns across multiple walking speeds, although the hip and knee joints tend to vary slightly more than the ankle joint between walking speeds[114]. At initial contact, the hip is in a flexed position and extends throughout most of the support phase before flexing in late support just prior to toe-off. This late-support flexion helps initiate the swing phase, during which the hip continually flexes until ground contact of the next step[27,114]. The knee is slightly flexed at ground contact and continues to flex to approximately 20° during weight acceptance, after which the knee extends during mid-support and then flexes rapidly to approximately 40° at toe off and continues to flex through the first half of the swing phase. During the final half of the swing phase, the knee extends to approximately 2° flexion immediately prior to heel contact of the next step[114]. At initial heel contact, the ankle plantarflexes until the foot is flat on the ground, at which point it dorsiflexes as the body's center of mass (COM) moves horizontally over and above the joint. During late support phase, rapid ankle plantarflexion causes push-off and aids in swing initiation[114]. These joint angular position curves remain similar across various walking speeds, however the minimum and maximum values as well as the slope of the change in position over time (joint angular velocity) do vary, creating an almost linear relationship between joint angular velocity and walking speed (with higher joint angular velocities at faster walking speeds)[114].

In biomechanics, kinetics refer to the forces and torques that cause movement of a body. Internal joint torques are used to quantify the net muscular forces acting across the joint at which these torques are being measured. During dynamic movements, torques cause rotations (changes in angular position) at skeletal joints. Joint powers represent the products of torques and angular velocities at each individual joint. Joint powers are particularly valuable in that they can be used to represent muscular contraction type, with positive power (energy generation) representing concentric (fiber-shortening) muscle contraction and negative power (energy absorption) representing eccentric (fiber-lengthening) muscle contraction. Joint work represents the change in energy at each joint and can be quantified by integrating the joint power-time curves. During concentric contractions, the muscles are doing work (supplying energy) to the skeleton. During eccentric contractions, muscles perform “negative work” and energy is subtracted from the system. In reality, during an eccentric contraction, the skeleton is performing work on muscles crossing a joint, however the “negative work” convention is prevalent in biomechanics and will be referred to throughout this thesis.

Healthy walking is considered a typical motor pattern in that muscle activation follows a proximal-to-distal sequencing pattern starting at the hip and moving distally to the knee and then the ankle. At the hip joint, a large extensor torque is present through the initial 35-40% of the support phase. During this initial portion of support, the hip joint is extending, first with a positive velocity and then a negative velocity. While the velocity is positive, a large positive power is being generated by the hip extensor muscles (i.e., gluteus maximus). As the hip continues to extend through the support phase, a flexor torque develops as the ground reaction force (GRF) vector moves anterior to the hip joint center. Here, the torque and angular velocity are in opposite directions and the joint power becomes negative, representing the eccentric action of the hip flexors as the leg continues to extend. At the knee joint, there is initially a slight flexor torque in the late swing phase and early support phase as the hamstrings act eccentrically while the knee extends, likely to prevent hyperextension of the knee joint just prior to ground contact. Immediately following this, during weight acceptance, an extensor torque is produced as the knee flexes to ~20°. Because the knee is continuing into flexion, the joint torque and velocity are in opposite directions, producing a negative power curve that represents the eccentric action of the knee extensor muscles. As the knee extends throughout mid-support, a flexor torque is produced. The knee then flexes again in late stance while the GRF vector moves posterior to the knee joint center, causing a slight extensor torque. At the ankle joint, there is an initial dorsiflexor torque following heel strike as the forefoot is lowered to the ground. Following this, a plantarflexor torque develops as the COM moves horizontally over the joint while the resultant GRF vector remains anterior to the joint center. At the end of support, the ankle rapidly plantarflexes, causing a large positive spike in the ankle joint power curve that represents concentric action of the ankle plantarflexor muscles. Quantifying joint-level kinetics allows for precise analysis of individual gait characteristics.

In healthy young adults, the net muscular torque is largest at the ankle joint and lower and more variable at the knee and hip joints[29]. The variation in knee and hip joint torques led to the proposal of a summed measure of all joint torques termed the “support” torque[113]. The support torque is a net extension pattern of joint torques at the hip, knee, and ankle acting to resist flexion of the lower limb joints caused by gravity and the GRF. Despite variations in more proximal joints (hip and knee), the support torque remains similar across healthy individuals walking over level ground[113]. This similarity remains even in pathological individuals, suggesting that if one joint lacks supporting capabilities, another joint may exhibit an increased contribution to overall support as a compensatory mechanism[113].

Similar to the consistency of joint angular velocity during walking discussed previously, joint power curve patterns remain similar across different walking speeds in healthy individuals, however minimum and maximum values, as well as the slopes of these curves, vary almost linearly with walking speed (with higher maximum values and increased slopes at faster speeds)[114]. In particular, plantarflexor power in late stance (referred to by D.A. Winter as the "power burst"[114]) assists in forward and upward acceleration of the push-off limb in order to initiate swing phase. An increased "power burst" results in a shorter swing period, a longer a step length, and thus a faster walking speed[114]. This mechanical finding has been confirmed with EMG data, showing increased activation of the ankle plantarflexor muscles at faster walking speeds[81,114].

Overall, the gait biomechanics discussed above suggest that level walking is a complex, multi-joint motor pattern governed, in healthy individuals, by a unique motor program that relies on proper neuromuscular timing (hip to knee to ankle). However, alternate gait patterns, such as incline walking gait, are also commonly used by most people every day. Therefore, these alternate gaits have also been analyzed.

Differences in joint kinematics and kinetics between level and incline walking show that ascending inclines requires its own unique motor program. With regards to kinematic differences, as healthy young adults ascend increasing inclines, both the hip and knee joints are in a more flexed position at initial contact and extend more throughout mid-support to raise the body's COM up the incline and flex to a greater extent throughout the swing phase to ensure toe clearance[70]. The ankle joint remains dorsiflexed for most of support phase but plantarflexes to the same degree as level walking during push-off in late support[70].

Joint level kinetic differences also exist between level and incline walking gaits in young adults. At the hip, early support phase joint torque and power increase during incline ascent compared to level walking[37,70]. EMG data for hip musculature support these mechanical findings. Specifically, EMG-measured muscular activity of the biceps femoris and gluteus maximus increase by 635% and 345% (relative to level walking), respectively, while ascending a 9° incline[40]. Although increased rectus femoris and vastus medialis EMG activity increased during incline ascent, mechanical findings have shown mixed results for knee joint torque and power between level and incline walking[37,40,70]. At the ankle, late support phase joint torque increases during incline ascent, however it appears to plateau at large inclines[70]. Interestingly, ankle joint power during 9° incline ascent was not significantly different than level walking[37]. However, EMG data showing increased medial gastrocnemius and soleus activity (175% and 136% relative to level, respectively) during 9° incline ascent suggests that ankle musculature activity does increase during incline ascent[40]. Overall, it appears that muscular activity at the hip, knee, and ankle joints increase during incline ascent compared to level walking.

Individual joint differences result in total lower limb differences between level and incline walking. Specifically, increased hip and ankle joint extension torques result in an increased support torque during incline compared to level walking[70]. Increased hip joint power, and to a lesser extent, increased ankle joint power, during incline ascent result in increased individual limb positive work compared to level walking[37,41]. In fact, each individual limb contributes only positive work during incline ascent[41]. This is true for even the leading leg during double support – which, in level walking, performs negative work during double support. Given that incline ascent requires significantly greater power generation (positive work) from each leg as well as increased muscular activity at each joint, it appears to be the more mechanically demanding task[40,41]. This is supported by the increased metabolic cost of incline walking[60], possibly due to decreased mechanical efficiency (less efficient transfer of potential and kinetic energy)[50] associated with incline walking. All of these findings conclude that incline ascent requires a motor pattern distinctly different from level walking and appears to be a more demanding task.

This sub-section detailed the kinematic and kinetic characteristics of healthy, young individuals during both level walking and incline ascent. These two modes of locomotion are each governed by their own unique motor pattern which can be quantified by joint-level biomechanical analyses.

Declining muscular function with age

Aging has been associated with decreased strength[49,67,118] and power[5,32,69] through several mechanisms. Specifically, loss of muscle mass and neuromuscular changes, such as loss of total motor units, that occur naturally with age have been linked to decreases in strength and power in healthy old adults and even successful masters athletes[30,75]. That even competitive old athletes exhibit muscular declines affirms that these changes occur naturally with age and do not necessarily appear to be pathological. Further, the muscles of the lower limb appear to decline at a faster rate than those of the upper extremities[15]. Age-associated loss of skeletal muscle mass and function has been termed sarcopenia[33]. It is estimated that 12% of people older than 60 years and 30% of people older than 80 years have sarcopenia[79]. Declining muscular function in old adults can have negative impacts on performance of even elementary daily activities. The increased rate of muscular decline in the lower extremities may have an even more pronounced impact on daily activities such walking. As an old individual loses the ability to perform activities as elementary as walking, he/she loses independence. Thus, it is not surprising that sarcopenia has been associated with functional decline and disability in old individulas[33]. This also places a strain on the U.S. economy, as sarcopenia and its associated disabilities add the cost of an estimated $18.4 billion per year to the U.S. health care system[58].

It is clear that age-related muscular decline is costly, not only to the U.S. economy, but perhaps more importantly, to the individuals who experience difficulty completing even elementary daily activities. An inability of old adults to complete daily activities reduces their overall quality of life, both socially and physically. Understanding how age and disability or low physical capacity may impact walking, an elementary function of daily life, may help guide training interventions aimed at improving quality of life in these populations. To accomplish this, a better understanding of the impact of aging and physical capacity on walking gait in old populations is needed.

Biomechanical plasticity in old adults

Level walking gait alterations have been observed between healthy old and young adult populations. Spatiotemporal variables such as decreased comfortable[115] and maximal[11] walking speed and decreased step length[59] compared to young adults are commonly reported in old adults. Walking speed decreases between 0.7%[10] and 1.2%[3] per year in adults over 60 years old. Decreased walking capacity in old adults may impair their ability to complete tasks of daily life and ultimately result in a decreased quality of life. Walking speed alone is a particularly important clinical variable as it is relatively easy to measure and is a reliable predictor of falls[1], disability[82], hospitalization[1], and even mortality[105] in old populations. Because of its apparent clinical importance, the causes of this age-related decrease in walking speed have been investigated.

As such, the underlying biomechanical differences between healthy young and old adults during level walking have been examined. The following paragraph describes kinematic differences reported in the literature between young and old adults. Decreased hip extension in late stance, possibly resulting from “tightness” of the hip flexor muscles (i.e., hip contracture), has been observed in old compared to young adults[64]. At the knee, reduced flexion range of motion during weight acceptance and reduced extension range of motion during mid-support have been observed in old compared to young[8]. Old adults also exhibit less ankle plantarflexion during push-off in the late support phase[59,78]. In fact, ankle plantarflexion motion in old adults during the transition from support to swing phase (push-off) is roughly 50% that of young adults[8]. During swing phase, lower ankle dorsiflexion and knee flexion have been reported, which could increase fall risk in old adults due to a decreased toe clearance[8]. These kinematic findings suggest that total hip, knee, and ankle joint ranges of motion are limited in healthy old compared to young adults during both support and swing phases of level walking.

Kinetic differences between healthy old and young populations have also been observed. Specifically, compared to young, old adults exhibit increased hip joint extensor torque and power during early support[24] and hip joint flexor power during late support[59]. Mixed results for knee joint differences between young and old have been reported. DeVita & Hortobagyi (2000) reported decreased angular impulse and work at the knee joint[24] during the initial loading response in early support while others have reported increased knee joint kinetics during this phase[78,115]. At the ankle joint, decreased torque and power have been reported in old compared to young adults during late support at self-selected[59,92,102,115], controlled[24], and fast[64,102] walking speeds. Differences in joint kinetic contributions between old and young adults were precisely quantified by DeVita & Hortobagyi (2000), who reported hip extension angular impulses of 17.1 Nms^{-1} in old adults compared to 10.8 Nms^{-1} in young adults[24]. At the ankle, the same authors reported plantarflexion angular impulses of 24.6 Nms^{-1} in old adults compared to 32.0 Nms^{-1} in young adults[24]. Additionally, healthy old adults produced 279% more work at the hip and 29% less work at the ankle compared to a younger cohort while walking at a similar speed (1.5 ms^{-1})[24]. All of these results indicate that old adults increase hip and decrease ankle joint contributions to gait during the support phase of level walking.

The distal-to-proximal shift from ankle to hip joint contributions to the support phase of level walking gait in old adults is one example of biomechanical plasticity and appears to occur naturally with age - that is, it is commonly observed in *healthy* old adults[24,59]. Biomechanical plasticity is a fundamental principle in the field of biomechanics that quantifies altered joint-level biomechanical function with altered states of health. In addition to the distal-to-proximal redistribution pattern of joint contributions in old adults, biomechanical plasticity has been observed in ACL injured adults[25,26] and morbidly obese adults both before and after weight loss[23,55]. These redistribution patterns coincide nicely with D.A. Winter's original proposal of the support torque, where he theorized that pathological individuals may alter or redistribute joint contributions in order to maintain a support torque similar to healthy individuals[113]. Although aging is not "pathological" per se, it is possible that even healthy old adults shift reliance to the muscles of the hip due to declining function of the muscles crossing the ankle joint. This redistribution theoretically allows old adults to maintain a support torque similar to young adults[24].

Because plantarflexion power contributes in multiple ways to walking, decreasing power capacity of the ankle joint has negative impacts on walking performance[81]. In fact, decreased plantarflexor power is a strong predictor of shortened step length[59], which contributes to overall walking speed. The reduction in ankle joint contribution to gait may be caused by the age-associated decline in muscular function noted specifically in the ankle plantarflexors[15,106,112]. In fact, Bendall et al. (1989) reported a significantly positive relationship between maximal triceps surae strength and walking velocity in old adults[10]. Silder et al. (2008) reported a significant positive correlation between maximal isokinetic plantarflexor strength and maximal plantarflexion power generated at both self-selected and fast (120% of self-selected speed) walking speeds in healthy old adults[102]. Old adults exhibit a distal-to-proximal redistribution of joint torques and powers, relying more heavily on hip, and in some cases knee, joint contributions to the support phase of level walking, likely to compensate for decreased muscular function of the ankle plantarflexors[24]. It is possible that differences in health status among old cohorts led to the mixed results for the knee joint. Although all studies discussed thus far have included healthy old adults, it is possible that some of these individuals exhibit decreased muscular capacity of the knee extensor muscles as well as the plantarflexor muscles and thus shift more reliance towards the hip extensors. The afore-mentioned increase in *late* support hip flexor power and decreased plantarflexor power immediately prior to swing phase in old adults suggests that this population adopts a “pull” strategy rather than the “push” strategy utilized by healthy younger adults for swing phase initiation[59]. That is, old adults may pull their leg into swing phase using their hip flexor muscles while young adults use their plantarflexor muscles to propel their leg into swing phase. Old adults also increase hip extensor power in the *early* support phase[24]. It is possible that the increased hip flexor power in late support noted by Judge et al. (1998) represents a heavier reliance on ipsilateral hip musculature for swing initiation and the increased extensor power noted by DeVita & Hortobagyi (2000) in early support represents work done by the contralateral leg to assist in forward propulsion of the body's COM. Taken together, it is possible that some old adults increase both hip extensor power during early support to aid in forward propulsion and flexor power during late support for swing phase initiation. Regardless, all of these results indicate the same thing: old adults exhibit an age-associated biomechanical plasticity represented by a distal-to-proximal redistribution of joint torques and powers during level walking.

Age-associated differences during incline ascent have also been observed. Findings from level walking in old adults demonstrate a decreased propulsive ability in this population. Because incline walking requires increased generation of positive power in order to propel the body up the incline, it is likely that old adults struggle to negotiate inclines. Accordingly, healthy old adults ascend inclines at slower speeds and with shorter step lengths compared to young adults[31]. Slower incline walking speed is due, at least in part, to an impaired ability for total trail leg power generation during incline ascent in old compared to young adults[38]. Observations of joint-level kinetics reveal that the same age-associated biomechanical plasticity observed during level walking also exists during incline ascent. Compared to young adults, old adults exhibit increased hip joint torque and power during late support and decreased ankle joint torque and power during late support (push off) of incline ascent[37,61]. In fact, for a 9° incline, old adults do not increase plantarflexion torque at all, suggesting its importance as a limiting factor for incline ascent performance[37]. These joint-level findings were observed while old and young adults walked at similar velocities (1.2 ms^{-1}) on an instrumented treadmill, eliminating any possible speed effects on joint kinetics. EMG data showing increased muscular activity in the gluteus maximus and decreased muscular activity in the medial gastrocnemius of old compared to young adults during incline walking confirms this mechanical-based evidence for biomechanical plasticity[39]. Increasing task difficulty places a greater load on the muscles of the lower limb, possibly causing a greater need for biomechanical plasticity in old adults. This is likely the case for incline walking, where healthy old adults do not increase plantarflexion power at all[37]. Increased hip extension and decreased knee extension and ankle plantarflexion torques in old compared to young adults have also been reported during stair ascent, another relatively difficult task for this population[61]. Additionally, some healthy old adults are not able to increase plantarflexion power while walking at a maximal speed over level ground[59]. However, Kerrigan and colleagues (1998) reported that their healthy old maintained the ability to increase plantarflexion power at faster compared to self-selected level walking speeds[64]. The age-related reduction in ankle joint plantarflexor torque and power generating capacity also becomes more pronounced during running and sprinting in old adults[66]. Overall, it appears possible that biomechanical plasticity increases in magnitude as task difficulty increases. It also seems possible that some task difficulty threshold exists at which point old adults can no longer increase ankle propulsive contributions to locomotion *at all* and thus, shift even more reliance to proximal joint contributions – this appears to be the case for incline ascent.

This sub-section has detailed the gait adaptations that occur in healthy old adults. To summarize, this population exhibits an age-associated biomechanical plasticity defined by a distal-to-proximal redistribution of joint contributions from the ankle to the hip. This shift exists not only during level but also during incline ascent. It appears that more difficult tasks, such as ascending inclines, climbing stairs, and in some cases increasing level walking speed, increase the magnitude of this age-associated biomechanical plasticity. These age-associated gait adaptations are likely due, in part, to declining muscular function, particularly of the distal muscles of the lower limbs, in even healthy old adults.

Biomechanical plasticity in low-capacity old adults

All of the studies reviewed thus far have reported differences observed between *healthy* old and young adult groups. However, not all adults age in a healthy and robust manner. Age-associated syndromes such as sarcopenia and frailty, as well as functional limitations and physical disabilities, may also impact walking capacity in old adults, regardless of chronological age. Experts on aging and its effects on skeletal muscle have suggested that old individuals with a normal walking speed slower than 1.0 m/s^{-1} should be assessed for sarcopenia[33]. Additionally, a walking speed slower than 0.8 ms^{-1} seems to be a reliable predictor of frailty in old adults[14]. Thus, it appears that decreased walking capacity may also occur as a function of declining physical capacity as well as increased chronological age. Indeed, compared to healthy old adults, *low-capacity* old adults exhibit slower walking speeds, shorter step lengths, and increased stride variability during level walking[21,71]. Interestingly, others have shown no difference in comfortable walking speed between healthy and low-capacity old adults[52,72]. McGibbon & Krebs (1999) did not control walking speed, however they did control step cadence during data collection, which most likely impacted walking speed of their participants[72]. The similarity in comfortable walking speed between healthy and low-capacity old adults reported by Graf et al. (2005) is surprising, particularly because they used a physical testing battery (part of which includes walking speed) to define physical capacity in their population[52]. The underlying joint-level biomechanics offer a more precise view of walking differences between healthy and low capacity old adults.

Joint-level kinematic differences between healthy and low-capacity old adults have been observed. At self-selected walking speeds, low-capacity old adults exhibit reduced hip extension[52] and ankle dorsiflexion[52] throughout support compared to healthy old adults. Others have observed reduced dorsiflexion in low-capacity old adults compared to healthy young adults, but failed to observe significant kinematic differences between low-capacity and healthy old adults[71]. Inconsistencies in kinematic results suggest the importance of comparing joint-level kinetic differences between healthy and low-capacity old adults.

Thus, joint-level kinetic differences between healthy and low-capacity old adults have also been examined. At the hip joint, increased peak extensor power during early support[52,71,72] and flexor torque and power in late support[71] have been observed in low-capacity compared to healthy old adults. At the knee, some differences between healthy young and low-capacity old adults have been observed, but differences between healthy and low-capacity old adults have yet to emerge in the literature[52,71]. At the ankle joint, decreased peak plantarflexor power and work during late support have been observed in low-capacity compared to healthy old adults[52,71,72]. These results are supported by Buddhadev & Martin (2016), who recently reported non-significant ($p < 0.10$) trends of increased relative hip and decreased relative ankle joint work in sedentary compared to active adults[13]. Joint-level kinetic differences between healthy and low-capacity old adults show that the same age-associated biomechanical plasticity occurs in both groups. However, further increases in hip and decreases in ankle joint contributions to support phase walking mechanics in low-capacity old adults suggest an increased magnitude of biomechanical plasticity compared to healthy, age-matched controls[52,71,72].

Although recurrent falls do not necessarily indicate low-capacity, differences between old fallers and non-fallers have been observed. Specifically, Kerrigan et al. (2000) showed joint-level kinetic differences between old fallers and non-fallers at both self-selected and “fast” walking speeds[63]. When comparing self-selected speeds between fallers and non-fallers, nine variables showed significant differences. However, when the fallers walked at a faster speed, only four variables remained significant: increased peak hip extensor torque during early support, decreased hip flexor torque during late support, decreased peak knee extension torque, and knee power absorption in pre-swing[63]. Decreased ankle positive power in pre-swing was also noted in the fallers, however this result was not reported as significant[63]. The lack of statistical significance stems from this group’s low *a priori* alpha level ($p < 0.0018$). Although these results offer insight into gait adaptations in old fallers, numerous methodological issues make comparison to the proposed study difficult. For example, “fallers” included by these authors were identified by falls history. It is possible that fallers exhibited gait adaptations in response to injury or biological disturbances that resulted from previous falls. A similar question answered using a prospective design might illuminate true biomechanical differences between fallers and non-fallers. However, if the assumption is made that old fallers exhibit lower physical capacities than the non-fallers, these results indicate increased hip joint contributions in old fallers, possibly in response to decreased ankle joint contributions.

Currently, literature exploring differences between healthy and low-capacity old adults during incline walking is not extensive. However, given the differences reported for level walking discussed previously, it is entirely possible, if not likely, that differences exist between healthy and low-capacity old adults. Stair ascent, which requires increased power generation from the joints of the lower limb[90], is commonly used as a performance measure for assessing physical capacity, with poorer performance associated with lower physical capacity[53]. Poorer performance of low-capacity old adults may arise from an inability to increase power from joints of the lower limb. Inability to increase joint power generation would limit incline ascent performance in low-capacity old adults. Graph et al. (2005) reported that low-capacity old adults maintained the ability to increase walking speed beyond the comfortable speed of their healthy old cohort and did this by increasing both hip extensor and ankle plantarflexor powers in early and late support, respectively[52]. However, peak plantarflexor power remained significantly lower in the low-capacity group walking at faster speeds (1.34 ± 2.55 ms^{-1}) compared to the healthy group walking at comfortable speeds (1.05 ± 1.65 ms^{-1})[52]. Thus, it appears likely that age-associated biomechanical plasticity becomes even further pronounced in low-capacity compared to healthy old adults during more difficult tasks, such as walking faster, ascending stairs, or walking up inclines.

Further muscular decline in the lower limbs of low-capacity compared to healthy old adults may account for the differences in walking ability between the two groups. That walking ability appears to be a valid predictor of sarcopenia[33], which is loss of muscle mass and function, supports this notion. Clark et al. (2013) divided a group of healthy old adults into “fast” and “slow” cohorts based on maximal walking speed[16]. Comparison of these two groups during a rapid plantarflexion movement showed decreased rate of force development in the plantaflexor muscles as well decreased rise of EMG signal in the medial gastrocnemius in the slow compared to fast group[16]. These results suggest that reduced muscular or neuromuscular function, particularly of the plantarflexors, may impact maximal walking speed in old adults. Also measuring healthy old individuals, Hortobagyi, et al. (2016) reported increased hip and decreased ankle joint relative contributions to total positive joint work in weak (48.5% and 38.7%, respectively) compared to strong (41.3% and 46.6%, respectively) old adults during level walking[56]. These authors measured total lower limb strength using a leg press testing protocol on a dynamometer with both isometric (with knee flexion angles of 15°, 45°, 90°) and isokinetic (at 30°/s and 60°/s) conditions[56]. Although the old adults included by Clark et al. (2013) and Hortobagyi et al. (2016) were reportedly healthy, it is likely that the “slow” and “weak” cohorts, respectively, included individuals of lower physical capacities compared to the “fast” and “strong” cohorts, respectively. Together, results from these two studies suggest that reduced muscular function limits walking capacity in old adults. Reduced muscular function of ankle plantarflexors in low-capacity old adults may cause a larger shift towards hip joint contributions, increasing the magnitude of age-associated biomechanical plasticity. Interestingly, however, increased hip joint angular impulse and work during early support have been reported in old adults after undergoing a 10-week lower-extremity power training protocol[9]. Increased hip joint contributions to gait following a power training protocol goes against the concept that increasing physical capacity (in this case, by training) decreases biomechanical plasticity magnitude.

The studies reviewed throughout this sub-section identified differences between *healthy* or *high-capacity* and *low-capacity* old adults. The results suggest that low-capacity old adults exhibit an increased magnitude of age-associated biomechanical plasticity compared to healthy old adults. However, all of these studies separated and compared participants in two discrete bins: “healthy” or “high-capacity” versus “low-capacity,” “strong” versus “weak,” and “fast” versus “slow.” Thus far, no study has examined biomechanical plasticity during level walking or incline ascent while treating physical capacity as a continuous rather than discrete variable nor has developed a quantitative description of the magnitude of biomechanical plasticity in relation to physical capacity. Treating physical capacity as a continuous rather than discrete variable would be a novel approach to measuring and describing gait adaptations in old adults.

Defining and measuring physical capacity

While the use of 3-D gait analysis and biomechanical analysis software is the standard for precisely and reliably quantifying lower limb joint kinematics and kinetics in biomechanics laboratories, tools for defining and measuring physical capacity are not so standard. Selecting a tool that accurately and reliably measures physical capacity is a necessity for the proposed analysis. Physical capacity is a measure of one's ability to perform activities commonly encountered in daily life. Tests such as the 36-Item Short-Form Health Survey (SF-36) and the Short Physical Performance Battery (SPPB) have been developed as a means of quickly and easily assessing physical capacity in old populations[53,110]. The SF-36 is a self-report questionnaire meant to survey health based on 8 broad categories, each with their own set of subcategories[110]. This questionnaire is easy to administer and has been deemed appropriate for use in determining health statuses across diverse populations[73]. The SPPB is a physical test of balance, gait, strength, and endurance by examining the ability to stand with feet in three different positions, time to walk 8 feet, and time to stand and sit in a chair 5 times[53]. The SPPB is a particularly useful protocol for the assessment of walking performance as it tests physical functionality of the lower-limbs. Results from this protocol have been shown to be independent predictors of disability in activities of daily life, worsening mobility, loss of ability to walk distances as short as 400 meters, nursing home admittance, and even mortality in old subjects[53,77,109]. These, coupled with other physical capacity or functionality testing procedures[18,108], provide relatively easy means of measuring physical capacity in old populations. Collins et al. (2004) reported a relationship between scores on their functional fitness testing protocol and the performance of activities of daily life, but not with age or chronic illnesses[18]. This reaffirms that, although some functional decline may occur naturally due to the aging process, the *rate* at which these declines occur vary between individuals. It also suggests that low scoring individuals may struggle with activities of daily life (i.e., walking), regardless of their chronological age or whether or not they have chronic illness. In these regards, it is possible to have low physical capacity without a chronic illness. For example, sarcopenia is not a chronic illness, but rather a health syndrome that may reduce the physical capacity of old adults.

Related to physical capacity is the concept of frailty, which has been loosely defined as a reduced homeostatic reserve and resiliency to stressors[95]. Fried et al (2001) defined a frailty phenotype as the presence of three or more of the following characteristics: unexplained weight loss, muscle weakness, self-reported exhaustion, slow walking velocity, and low activity level[46]. Although it is commonly associated with impairment, disability, and aging, frailty has been shown to be distinctly different from both chronological age and disability[45,94]. In fact, 98% of surveyed geriatricians (n = 62) reported that frailty is a syndrome of its own[45]. In reviewing the literature, Rockwood et al. (2000) concluded that frailty is effectively biological rather than chronological aging[94]. To further differentiate frailty from disability, the same authors concluded that frailty arises from multisystem dysfunction whereas disability may arise from the dysfunction of a single system and that frailty is necessarily associated with instability whereas disability is not[94]. In this case, instability was defined as the disproportionately large effects of small environmental perturbations[94]. Like reduced physical capacity, frailty has been associated with adverse outcomes[14]. Therefore, even without a clear and concise definition for frailty[34], obtaining a proper measurement could prove valuable in studies including old adults.

Methods for measuring frailty vary from study to study, most likely due to the varying definitions for frailty. The phenotype proposed by Fried et al. is commonly used to determine whether populations are robust, pre-frail, or frail[46]. This method allows for categorical groups to be formed within study samples, but is not an efficient way of setting a wide range of frailty values. Because a range of physical capacities is often more useful, other, more robust methods have been developed in an attempt to create a more accurate severity scale for frailty. Rockwood et al. (2005) developed the Frailty Index (FI), a list of 70 deficits including the presence and/or severity of current diseases, ability in activities of daily living, and physical signs from clinical and neurological examinations[96]. When applying the FI to study participants, a 70-dimensional vector is constructed to calculate an index score (for example, seven deficits would yield 7/70 = 0.10) which serves as that individual's level of frailty[96]. Measuring frailty using the FI has been cross validated and appears to be more precise than the Fried et al. phenotype in determining the risk for adverse outcomes in the old populations[93]. This method would allow for variables, such as biomechanical gait alterations, to be mapped across a range of frailty scores rather than compared between the 3 distinct groups formed using the Fried et al. phenotypic method of defining frailty (robust, pre-frail, frail). Additionally, the FI has been shown to count as frail some subjects that would fall under the pre-frail category of the phenotypic model, suggesting that the F.I. may allow for more sensitive and accurate measurement[103]. Thus, it appears that using a more precise model, such as the FI, yields better results. However, because the FI incorporates 70 variables, it is often far more time consuming. To reduce test completion time, Rockwood et al. also developed the Clinical Frailty Scale, which ranges from 1 (robust/healthy) to 7 (complete functional dependence on others) [96]. While a 1-7 range is better than 3 distinct groups (robust, pre-frail, frail), it is still a very small range and would not be appropriate for use in correlation and regression analyses. The Comprehensive Geriatric Assessment (CGA) was developed as a means of incorporating medical, functional, and psychological capacity in order to develop more specific treatment protocols for old adults of all capacities[98]. The CGA is considered by some to be the gold standard in detecting frailty, however the expertise required to perform the assessment limits its use in academic settings[17]. Multiple methods have been employed in detecting and measuring frailty and its associated adverse outcomes. This makes choosing a specific method rather difficult, however Rockwood et al. advises embracing the complexity of frailty and choosing a method/definition of frailty that has been proven and is most useful in the given situation[93,94].

It is important to note that it is possible to be completely healthy on these “frailty scales” (i.e., robust). “Frail” is the absolute end stage in the tests mentioned above. This stage represents the population of old adults with the lowest physical capacity or functional status. Logically an inverse relationship exists between frailty and physical capacity: as one becomes increasingly frail, his/her physical capacity declines. Frailty is a dynamic process, making transitions from different frailty states possible (i.e., increasing or decreasing physical capacity)[48]. Lang et al. described this dynamic process as the decrease in physiological reserves while increasing physiological resources are required to repair (in the case of a perturbations) and maintain proper function[68]. While transitions to greater frailty levels are more common, it is possible to transition into lower levels of frailty (i.e., increase physical capacity)[48]. This finding provides hope that training interventions, if developed and implemented properly, could be of great benefit in old populations of decreasing physical capacities. To this end, understanding the effects of physical capacity on walking gait may be beneficial in developing such intervention programs. This is particularly important given the overall benefit of daily walking in old adults[84,97]. Understanding gait adaptations in relation to physical capacity in old adults may help guide future attempts in developing proper training interventions for this population. Hopefully, these interventions will allow individuals in this population to increase their ability to walk, lessening their dependence for outside help and allowing them to reap the health benefits associated with regular physical activity.

The few studies that have made precise biomechanical comparisons between healthy and low-capacity old adults have employed various methods in order to define and measure physical capacity. Graf et al. defined individuals as “low-performance” if they scored 9 or less (out of 12 possible) on the SPBB[52]. McGibbon & Krebs (1999, 2004) defined individuals as “low-functioning” if they exhibited one or more physical limitations as defined by the SF-36 physical function scale[71,72]. Kerrigan et al. (2000, 2001) compared old fallers and non-fallers, however, falls don’t necessarily indicate low physical capacity[62,63]. Buddhadev & Martin (2016) separated sedentary and active old adults based on weekly physical activity time (min/week)[13]. Although they only included healthy individuals, Hortobagyi et al. (2016) used maximal leg press strength to separate and compare “weak” and “strong” old adults[56]. The use of different measures of physical capacity in these biomechanical studies demonstrates the difficulty in choosing the proper method for determining physical capacity in old adults.

The information reviewed in this sub-section identified tools commonly used for measuring physical capacity in old adults. All of these tools have strengths and weaknesses. For the proposed study, it is important to select a tool that accurately and reliably measures physical capacity, is relatively easy to conduct, and will produce scores across a range large enough to appropriately conduct correlation and regression analyses.

Walking gait alterations with decreasing physical capacity

Biomechanical differences between healthy and low-capacity old adults were discussed previously. This sub-section identifies the limited data available on gait differences observed across a range of physical capacities (i.e., studies treating physical capacity as a continuous rather than discrete variable). The majority of the findings reported here come from assessment techniques used to measure physical capacity or frailty in old populations, where poorer walking performance leads to poorer scores (lower physical capacities or increased frailty).

Among the characteristics commonly used to assess physical capacity and frailty, walking speed appears to be one of the more important variables. A recent review showed that self-selected walking speed alone can be a powerful and reliable test to predict complications associated with frailty in old adults[85]. Castell et al. (2013) suggests that a walking speed slower than 0.8 m/s is a simple and effective method of identifying frailty in old populations[14]. Additionally, walking speed appears to decline as individuals move from non-frail into pre-frail groups[44]. Slow walking speed is also a variable commonly used in tests that assess physical capacity[47,53]. Indeed, walking speed in adults over 65 years old appears to be a valid predictor of both physical and cognitive function[47]. Other spatiotemporal variables such as shorter step length, decreased single-support time, and increased double-support time have also been associated with increasing frailty levels (pre-frail versus non-frail)[44]. Overall, it appears that walking performance decreases as physical capacity declines in old adults. One possible mechanism of decreased walking speed in low-capacity old adults is sarcopenia, which has been associated with both pre-frail and frail old adults[83]. Logically, physical capacity will decrease as individuals lose muscular capacity and thus, their ability to perform tasks of daily life (such as walking) will decrease.

Currently, there are limited data relating physical capacity to incline walking performance. However, it is worth noting that some assessment tests for physical capacity include stair ascent performance measures[53]. For example, scores on the stair climb power test

(SCPT) can be used to accurately predict mobility impairments in old adults[7]. Stair ascent is similar to incline walking in that, compared to level walking, the hip, knee, and ankle joints must produce more positive power throughout the majority of the support phase[90]. Thus, difficulties ascending stairs most likely indicate difficulties negotiating inclines. In tests measuring physical function or capacity in old adults, difficulty completing these tasks (stair or incline ascent) would negatively impact scores and indicate lower physical capacity. Thus, it appears likely that, as physical capacity declines, the ability of old adults to ascend inclines also declines.

Given the clinical importance of decreased walking speed[1,82,105] , the apparent direct relationship between physical capacity and walking speed warrants further investigation. Specifically, understanding the joint-level biomechanics of level and incline walking gait while treating physical capacity in old adults as a continuous rather than discrete variable is an important next step in the biomechanical literature. The results will provide joint-specific adaptations to declining physical capacities and could prove useful in the future development of training interventions aimed at this population. However, no study has investigated biomechanical alterations in this context.

Hypothesis & purpose

Biomechanical differences between healthy old and young adults as well as between healthy and low-capacity old adults are established in the literature. These studies quantified joint-level biomechanical differences between these populations (healthy young versus healthy old and healthy or high-capacity versus low-capacity old). Currently, no study has treated physical capacity as a continuous rather than discrete variable, across which biomechanical alterations may exist. Given the information reviewed here, it is hypothesized that age-associated biomechanical plasticity becomes more pronounced as physical capacity declines in old adults.

More specifically, as physical capacity declines, old adults will rely more heavily on hip joint contributions and less on ankle joint contributions during the support phase of walking. It is also hypothesized that the magnitude of biomechanical plasticity becomes more pronounced during the more challenging task of incline walking. In other words, the increase in magnitude of biomechanical plasticity per unit change of physical capacity will be higher during incline compared to level walking. These adaptations will be quantified by hip, knee, and ankle joint torques, angular impulses, powers, and relative work in relation to physical capacity scores measured using the SF-36 and 20-meter self-selected walking speed. The purpose of this thesis is to examine the relationships between physical capacity and age-associated biomechanical plasticity during level and incline walking. The results will precisely and mathematically quantify the relationships between physical capacity and biomechanical plasticity during both level and incline walking.

Summary

Both healthy and pathological aging are associated with decreased skeletal muscle mass[49], strength[49,86], and power[5,32]. Declining muscular function in old adults can have a profound impact on this population's ability to perform daily activities such as walking and can result in a reduced quality of life. Indeed, reduced walking capacity in old adults, including shorter step length[115] and slower walking speed[11], is predicative of falls, disability, hospitalization, and even mortality[1,82,105]. Because of the importance of decreased muscle quality and subsequent decline in walking performance, the underlying biomechanical components of walking gait in old adults have been examined.

Compared to young adults, healthy old adults exhibit increased hip joint extensor contributions during the early support phase and decreased ankle joint plantarflexor contributions during the late support phase of level walking at both self-selected[115] and controlled[24] speeds. These biomechanical differences were quantified by DeVita & Hortobagyi (2000), who reported that, while walking at the same speed (1.5 ms^{-1}), hip and ankle joint contributions to total positive joint work were 44% and 51% respectively for old adults and 16% and 73% respectively for young adults[24]. That represents a 279% increase in work done at the hip and a 29% reduction in work done at the ankle in old compared to young adults[24]. Redistribution of joint contributions is termed biomechanical plasticity and is considered a fundamental biomechanical principle that quantifies altered joint-level biomechanical function with altered states of health. In addition to the distal-to-proximal redistribution pattern of joint contributions in old adults, biomechanical plasticity has been observed in ACL injured adults[25,26] and morbidly obese adults both before and after weight loss[23,55]. Age-associated biomechanical plasticity has also been observed during incline ascent, where old adults exhibit increased hip and decreased ankle joint contributions to the support phase compared to young adults[37]. However, at a 9° incline, old adults do not increase ankle joint plantarflexor torque at all compared to level walking, suggesting a more pronounced biomechanical plasticity during this more challenging task[37]. Decreased muscular function, particularly of the ankle plantarflexors[69], in old adults is likely a major component of this age-associated biomechanical plasticity during locomotion. Indeed, Hortobagyi, et al. (2016) reported increased hip and decreased ankle joint relative contributions to total positive joint work in weak (48.5% and 38.7%, respectively) compared to strong (41.3% and 46.6%, respectively) old adults during level walking[56]. More difficult tasks, such as ascending inclines, that place a greater load on the muscles of old individuals relative to their functional capacity, may cause biomechanical plasticity to become more pronounced.

The previously discussed gait adaptations pertain to *healthy* old adults. However, not all adults age in a healthy and robust manner. To determine possible effects of physical capacity on gait adaptations, comparisons have been made between healthy or *high-capacity* and *low-capacity* old adults. Compared to healthy old, low-capacity old adults walk with slower gait speeds[71], the importance of which was discussed previously. Joint-level biomechanical comparisons between high-capacity and low-capacity old adults show that the same age-associated biomechanical plasticity exists in both groups, however the magnitude of plasticity appears to be larger in the low-capacity group[13,52,71,72]. More specifically, increased peak hip joint extensor power during early support[52,71,72] and flexor torque and power in late support[71] as well as decreased peak ankle joint plantaflexor power and work[52,71,72] in late support have been observed in low-capacity compared to high-capacity old adults. Currently, no study has directly compared high-capacity and low-capacity old adults during incline walking. However, given that incline walking requires increased power generation from both the hip and ankle joints[37], and the apparent lack of power generation capacity in the ankle joint of low-capacity old adults, it is likely that the low-capacity group will rely on an even larger shift towards hip and away from ankle joint contributions compared to high-capacity old adults during incline walking. Further decline of muscular function in low-capacity compared to high-capacity old adults is likely a major component of the increased magnitude of plasticity in the low-capacity groups. All of these results indicate that low-capacity old adults exhibit larger magnitudes of age-associated biomechanical plasticity compared to high-capacity old adults, likely due, at least in part, to further muscular decline. However, all of these studies separated high and low-capacity old adults into two discrete bins for between-group comparisons. Thus far, no study has examined biomechanical plasticity while treating physical capacity as a continuous rather than discrete variable, across which biomechanical adaptations may occur.

Based on the literature reviewed throughout this chapter, it is hypothesized that physical capacity and biomechanical plasticity are inversely related such that, as physical capacity declines, the magnitude of age-associated biomechanical plasticity increases. More specifically, as physical capacity declines, old adults will exhibit increased hip and decreased ankle joint contributions, quantified by joint torques, angular impulses, powers, and relative work, to support phase mechanics of walking. It is also hypothesized that the magnitude of biomechanical plasticity becomes more pronounced during the more challenging task of incline walking. In other words, the increase in magnitude of biomechanical plasticity per unit change of physical capacity will be higher during incline compared to level walking. The purpose of this thesis is to examine the relationships between physical capacity and age-associated biomechanical plasticity during level and incline walking. The results will precisely and mathematically quantify the magnitude of age-associated biomechanical plasticity with relation to physical capacity in old adults during both level and incline walking.

Chapter III: Methodology

Introduction

Based on the reviewed literature, two hypotheses were formulated: first, that as physical capacity declines in old adults, age-associated biomechanical plasticity increases in magnitude; second, that the magnitude of biomechanical plasticity becomes more pronounced during the more challenging task of incline walking. To test these ideas, we performed level and incline gait analyses on 10 healthy young adults and 32 old adults exhibiting a range of physical capacities and derived hip, knee, and ankle joint peak torques, angular impulses, peak powers, and work. Biomechanical plasticity, assessed from these kinetic variables, were correlated and regressed onto physical capacity scores of old participants to determine the relationships between physical capacity and age-associated biomechanical plasticity during level and incline walking. This section provides a detailed summary of the participant characteristics, inclusion/exclusion criteria, instruments, procedures, data analysis, and statistical analysis used to test our hypothesis.

Participant characteristics

Participants were recruited from Greenville, NC and the surrounding area. Recruitment methods included fliers placed in local establishments, newspaper ads, in-person recruiting at local recreation centers, and classroom announcements. This study included 10 young (7 female; age = 20.3 ± 1.5) and 32 old (22 female; age = 74.7 ± 4.4) adults. All participants were initially screened using a short health history form via telephone or in person to determine their eligibility for participation in this study (Appendix D). Participant characteristics are presented in **Table 1**.

All participants provided written informed consent (Appendix B) approved by the Institutional Review Board of East Carolina University (Appendix A).

Table 1: Mean and standard deviation values of participant characteristics.

Participant Characteristics				
	Young		Old	
Variable	**Mean**	**SD**	**Mean**	**SD**
Age (yrs)	20.3	1.5	74.7	4.4
Height (m)	1.74	0.09	1.67	0.07
Mass (kg)	69.0	12.9	68.4	11.7
BMI (kg/m^2)	22.7	2.5	24.5	3.4

Inclusion criteria for young participants

1. Aged 18 – 25 years old.
2. Apparently healthy with no lower limb musculoskeletal injuries or neuromuscular pathologies that may impair walking gait.
3. BMI less than 30 kg/m^2 to account for obesity effects on gait.
4. Moderately active – regularly participates in some form of physical activity (at least 3 times per week).
5. Provide written informed consent.

Exclusion criteria for young participants

1. Current use of tobacco products.
2. Any cardiovascular or neurological pathology.
3. Minor lower limb musculoskeletal injury or disorder in the previous 6 months.
4. History of lower limb or back surgery.

Inclusion criteria for old participants

1. Aged 70 – 85 years old.
2. Must not be homebound as all testing will take place in the Biomechanics Laboratory at East Carolina University.
3. Have the capacity to walk on level and inclined surfaces without assistance. This includes aid from canes, walkers, and other individuals.
4. BMI less than 30 kg/m^2 to account for obesity effects on gait.
5. Provide written informed consent.

Exclusion criteria for old participants

1. Current use of tobacco products.
2. Inability to complete the following daily activities without aid: walk, climb stairs, rise from a chair, dress oneself.
3. Lower limb joint replacement.
4. Terminal illness.

Instruments

A short health history questionnaire (Appendix D) was used to determine participant eligibility. During an initial laboratory visit, the Short Form Healthy Survey (SF-36) (Appendix C) was used to determine physical capacity of all participants[110]. Also during the initial visit, an electronic timer (TracTronix Wireless Timing Systems, Lenexa, KS) was used to measure self-selected and safe-maximal gait speeds over a 20-meter level walkway as well as to time the 6-minute walk test while an infrared timing system (TracTronix Wireless Timing Systems, Lenexa, KS) with timing gates placed 3-meters apart was used to measure self-selected incline walking speed. For biomechanical assessment during the second and final visit, kinematic data for both level and incline walking were collected using an 8-camera (ProReflex MCU 240) motion capture system (Qualisys AB, Göteburg, Sweden). Each camera was set at a capture frequency of 120 Hz. Ground reaction force data were collected simultaneously using force platforms (AMTI, Newton, MA). For level walking trials, an embedded force platform (AMTI Model BP6001200-2k, Watertown, MA) located in the middle of a 25-meter level walkway was used to capture ground reaction forces. For incline walking trials, a smaller force platform (AMTI Model OR6-6-2000, Watertown, MA) placed in the center of a constructed incline ramp (3.2 m long, 10°) was used to capture ground reaction forces. For incline trials, the force platform was located far enough up the ramp to allow for participants to take at least one step with the right leg (the collection leg) prior to contacting the force platform with the collection leg. The collection frequency of both force platforms were set at 960 Hz and a gain of 4000. For both force platforms, six analog channels were used to measure 3-D forces and torques acting on the platforms. Gait speeds during both level and incline walking trials were measured with an infrared timing system (TracTronix Wireless Timing Systems, Lenexa, KS) with timing gates placed 3-meters apart within the motion capture area. All data were collected using Qualisys Track Manager Software (Qualisys AB, Göteborg, Sweden) and analyzed using Visual 3D (C-Motion, Germantown, MD) and proprietary Lab software (QuickBasic).

Procedures

All testing was conducted in the Biomechanics Laboratory (332 Ward Sports Medicine Building) of East Carolina University, Greenville, North Carolina. A brief health history questionnaire was used as a screening tool to ensure participant eligibility. For all accepted participants, testing was spaced over two days. Upon arrival on the first day of testing, all participants were instructed to read and sign the informed consent form approved by the Institutional Review Board of East Carolina University.

During the initial Lab visit, all participants were instructed to complete the Short-Form Health Survey (SF-36) in order to quantify physical capacity[110]. The SF-36 is a self-report survey consisting of 36 questions regarding overall physical and mental health and function. For the SF-36, question responses are not open-ended; participants are given a list of possible responses from which to choose. The creators of the SF-36 have an online scoring mechanism that was used to calculate 8 sub-scores: 1) limitations in physical activities due to health problems; 2) limitations in social activities due to physical or emotional problems; 3) limitations in usual role activities due to physical health problems; 4) bodily pain; 5) general mental health (psychological stress and well-being); 6) limitations in usual role activities due to emotional problems; 7) vitality (energy and fatigue); 8) general health perceptions. The SF-36 also provides a physical component score and a mental component score. This survey has been deemed appropriate for use in determining health statuses across diverse populations[73]. The SF-36 accurately measures physical capacity, is relatively easy to conduct, and can produce scores across a range large enough to appropriately conduct correlation and regression analysis. Also during the initial visit, participants completed four hand-timed trials over a 20-meter level walkway. The 20-meter walkway was located in a long, unobstructed corridor and marked at the beginning and end by pieces of tape. All participants started a few strides behind the beginning mark and walked a few strides past the end mark to account for any possible acceleration or deceleration effects. A member of the research team followed the participant from behind while measuring time with an electronic timing device. For the first two of these trials, participants were instructed to walk at a self-selected walking speed. For the final two trials, participants were instructed to walk at a safe-maximal walking speed. The two-trial averages of these walking speeds served as additional measures of physical capacity[14,85]. A 6-minute walking test was also conducted during the initial visit. For this test, small cones were placed 100-feet along a straight line from each other in a long, flat, unobstructed corridor. Participants were instructed to walk back and forth between the two cones, turning around at each cone, for 6-minutes at a self-selected, comfortable walking speed. A member of the research team counted the number of "laps" completed and measured the distance covered over the 6-minute time limit. The member of the research team conducting this test did not follow or verbally interact with the participant at all during the 6-minute testing period in order to avoid any possible alteration in the participant's walking speed (i.e., motivational effects)[4]. The distance covered over the course of the 6-minute walking test was also used as a measure of physical capacity[4,28,54,87,91]. Finally, each participant completed two incline ascent trials at self-selected speeds up the incline ramp. The two-trial average of these walking speeds was used as a measure of physical capacity[14,85]. Immediately prior to these two trials, participants were allowed 2-3 practice trials in order to become familiar with the ramp. This concluded testing for the initial visit.

Upon arrival on the second day of testing, all participants were re-informed of the gait analysis protocol prior to beginning data collection. Spherical reflective markers were used to define the pelvis and segments of the right lower limb of each participant. The pelvis was defined using the right and left iliac crests and the right and left greater trochanters. The right thigh was defined using the right and left greater trochanters and medial and lateral femoral epicondyles of the right leg. The right leg was defined using the medial and lateral femoral epicondyles and the medial and lateral malleoli of the right leg. The right foot was defined using the medial and lateral malleoli and the 1st and 5th metatarsal heads of the right leg and foot. To capture segment motion during dynamic trials, 4-marker rigid plastic shells were placed on the lateral aspect of both the right thigh and leg while a 3-marker shell was placed on top of the right foot. Individual markers were also be placed on the right and left anterior (ASIS) and posterior (PSIS) superior iliac spines. Following a 5 second static calibration trial, the segment defining markers were removed while the motion tracking markers remained for the entire testing session.

Participants then completed walking trials under 4 separate conditions: level walking at a self-selected speed, level walking at a controlled speed (1.3 ms^{-1} ± 5%), incline walking (10°) at a self-selected speed, and incline (10°) walking at a controlled speed (1.2 ms^{-1} ± 5%). For self-selected speed trials, participants were instructed to walk as if they were "going to an appointment." Although condition randomization is ideal, level walking conditions were performed first for all participants in order to avoid any possible fatigue-effects from completion of incline walking conditions, particularly in our old cohort. Following completion of level walking conditions, the subjects were allowed a brief period of rest while the incline ramp was constructed on the Laboratory walkway (approximately 10 minutes). All participants then completed the incline walking conditions. For each condition, 5 successful trials were collected (for a total of 20 successful trials) for most individuals, however some old individuals were not able to continue testing long enough for all 5 trials to be completed. Trials were considered successful in all conditions if full right foot contact was made with the force platform and gait speed was within the accepted ranges. Unsuccessful trials were discarded and experimentation continued until 5 apparently successful trials were collected. To ensure full contact and avoid "targeting," the starting foot position of the participant was monitored and altered accordingly by a member of the research team. Also to avoid "targeting" and ensure a natural walking pattern, all participants were verbally instructed to look straight ahead and to walk as naturally as possible. Verbal feedback and instructions were provided regarding gait speed (i.e., “speed up” or “slow down”) when appropriate.

Data analysis

Data were collected using Qualysis Track Manager Software which produces and integrates marker position data and ground reaction forces. Data were then processed using Visual 3D. A subject-specific linked rigid-segment model of the pelvis and right lower limb was created using the static calibration trial taken at the beginning of the gait analysis. The static calibration trial was also used to locate virtual joint centers, each segment’s COM, and to define the local coordinate system of each segment as well as to determine the location of each individual reflective marker within the global coordinate system. The hip joint center was determined by calculating 25% of the distance between the right and left greater trochanter calibration markers while knee and ankle joint centers were determined by calculating 50% of the distance between the medial and lateral femoral epicondyle and medial and lateral malleoli calibration makers, respectively. A line from the distal to proximal virtual joint centers of each segment defined each segment’s longitudinal axis. Anthropometrics were used to determine each segment’s COM position.

Second order low-pass Butterworth digital filters with cut-off frequencies of 6 Hz and 45 Hz were applied to position and GRF data, respectively. These digital filters were used to remove high frequency noise from the data. The filtering process is particularly important for position data as the error becomes more and more pronounced when velocity (the first derivative of position with respect to time) and acceleration (the second derivative of position with respect to time) are derived from the position data. Clean acceleration data in particular is crucial for calculating joint torques using an inverse dynamics approach.

Visual 3D software uses linear and angular Newtonian equations of motion to calculate joint reaction forces (JRF) and joint torques using an inverse dynamics approach. The inverse dynamics approach uses ground reaction forces, center of pressure, segmental anthropometrics, and kinematic position and acceleration data for these calculations. The process begins with the foot, as that is the segment in contact with the force platform, and thus, the known ground reaction forces, and moves proximally to the leg and then the thigh. **Figure 1** provides a schematic of Free Body Diagrams (FBD) for each segment (thigh, leg, and foot). For joint torques, Visual 3D always uses the right hand rule in determining the sign (positive/negative) of the calculated torque.

The basis for all joint force and torque calculations, respectively, are the following:

$$\Sigma F = ma$$

$$\Sigma T = I\alpha$$

Where ΣF represents net force, m represents mass, a represents linear acceleration, ΣT represents net torque, I represents moment of inertia, and α represents angular acceleration.

The following equation is used to calculate the vertical ankle JRF ($Ankle_z$):

$$GRF_z + F_{COMF} + Ankle_z = (m_{foot})(a_{footz})$$

Where GRF_z is the vertical ground reaction force, F_{COMF} is the weight of the foot (the product of the mass of the foot and gravity), $Ankle_z$ is the vertical ankle JRF, m_{foot} is the mass of the foot, and a_{footz} is the vertical acceleration of the foot segment.

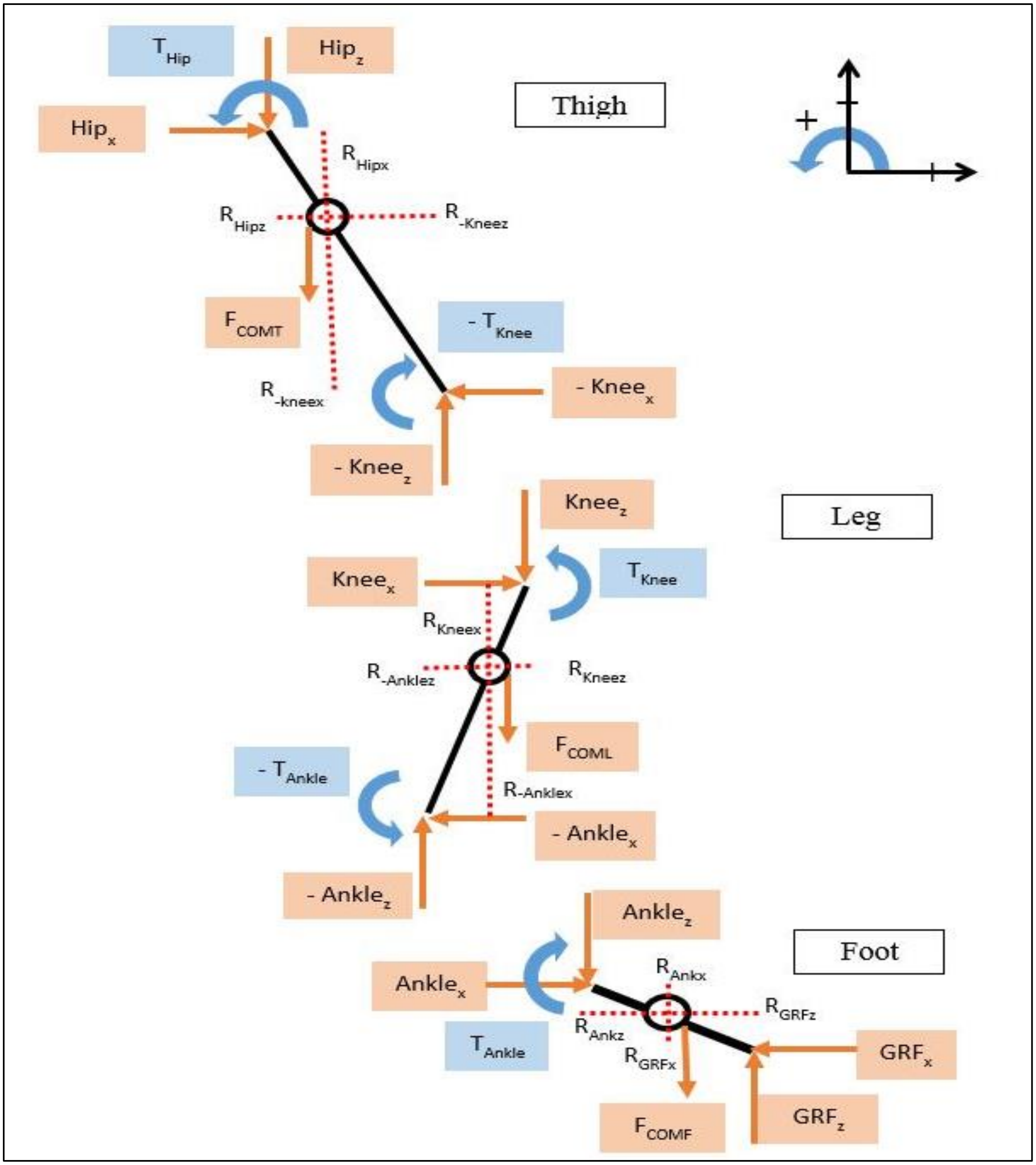

Figure 1: Schematic of free body diagrams for the thigh, leg, and foot segments used to calculate hip, knee, and ankle joint forces and torques. All vertical and horizontal arrows represent forces; all angular arrows represent joint torques; all circles represent segment COM locations; all dotted lines represent moment arms from the joint COM to each of the external forces.

The following equation is used to calculate the horizontal ankle JRF ($Ankle_x$):

$$GRF_x + Ankle_x = (m_{foot})(a_{footx})$$

Where GRF_x is the horizontal ground reaction force, $Ankle_x$ is the horizontal ankle JRF, and a_{footx} is the horizontal acceleration of the foot segment.

All external forces are then used to calculate joint torques. At the ankle joint, the following equation is used to calculate the joint torque:

$$GRF_z(R_{GRFz}) + GRF_x(R_{GRFx}) + Ankle_z(R_{Anklez}) + Ankle_x(R_{Anklex}) + T_{Ankle} = (I_{foot})(\alpha_{foot})$$

Where R represents the moment arms of each of the forces, I_{foot} is the moment of inertia of the foot, and α_{foot} is the angular acceleration of the foot. Center of pressure data for the GRF is essential in calculating moment arms for the vertical and horizontal components of the GRF. The center of pressure data of the GRF is measured by the force platforms. Linear and angular segment accelerations are derived from position data.

Inverse dynamics utilizes the JRFs and torque at the ankle to calculate knee JRFs and torque. The same general equations used for the ankle joint are also employed for the knee joint. As such, the following equations are used to calculate vertical and horizontal knee JRFs and knee joint torque, respectively:

$$-Ankle_z + F_{COML} + Knee_z = (m_{leg})(a_{legz})$$

$$-Ankle_x + Knee_x = (m_{leg})(a_{legx})$$

$$-Ankle_z(R_{-Anklez}) + (-Ankle_x)(R_{-Anklex}) + (Knee_z)(R_{Kneez}) + (Knee_x)(R_{Kneex}) - T_{Ankle} + T_{Knee} = (I_{leg})(\alpha_{leg})$$

The following equations are then used to calculate vertical and horizontal hip JRFs and hip joint torque, respectfully:

$$-Knee_z + F_{COMT} + Hip_z = (m_{thigh})(a_{thighz})$$

$$-Knee_x + Hip_x = (m_{thigh})(a_{thighx})$$

$$-Knee_z(R_{-Kneez}) + (-Knee_x)(R_{-Kneex}) + (Hip_z)(R_{Hipz}) + (Hip_x)(R_{Hipx}) - T_{Knee} + T_{Hip} = (I_{thigh})(\alpha_{thigh})$$

Joint powers are calculated as the product of each joint's torque and angular velocity. The following equation was used:

$$\vec{P} = \overrightarrow{JT}\ x\ (\vec{\omega}_{Proximal} - \vec{\omega}_{Distal})$$

Where P is a vector representing joint power, JT represents the XYZ components of the joint torque, and $\omega_{Proximal}$ and ω_{Distal} represent the proximal and distal segment angular velocities. Joint angular velocities are derived from segment position data. Finally, total positive and negative work, representing the sum of the positive and negative work, respectively, calculated at each of the three joints will be derived. All variables from the joint torques and powers were computed using proprietary Lab software. All joint kinetic variables are reported in the sagittal plane of motion. To define biomechanical plasticity, ratios of peak hip extensor torque to peak ankle plantaflexor torque, hip extensor angular impulse to ankle plantarflexor angular impulse, peak hip extensor positive power to peak ankle plantarflexor positive power, and hip extensor positive work to ankle plantarflexor positive work were computed.

Statistical analysis

Student's t-tests ($p < 0.05$) were conducted to determine differences in hip, knee, and ankle joint peak torques, angular impulses, peak positive powers, and work as well as differences in biomechanical plasticity ratios (described above) between young and old adults during all four conditions (these analyses were used to ensure that age-associated biomechanical plasticity existed within our study sample). Linear regression and correlation analyses were then conducted solely within the old group to determine relationships between our measures of physical capacity and biomechanical plasticity. Physical capacity was defined separately by each participant's score on the SF-36 and 20-meter self-selected walking speed collected during his/her initial visit. Pearson's correlation coefficients were computed between these two physical capacity measures and select joint kinetic variables as well as biomechanical plasticity ratios. All relationships were tested at the alpha = 0.05 level of significance. To test our second hypothesis, we compared level and incline walking at self-selected speeds and level and incline walking at controlled speeds. In the cases where corresponding significant correlation coefficients existed, we computed 95% confidence intervals of the regression slopes (beta weights).

Chapter IV: Results

Introduction

The purpose of this study was to examine the relationships between physical capacity and age-associated biomechanical plasticity during level and incline walking. Based on previous literature it was hypothesized that, as physical capacity declines in old adults, age-associated biomechanical plasticity becomes more pronounced. It was also hypothesized that the magnitude of biomechanical plasticity becomes more pronounced during the more challenging task of incline walking. In other words, the increase in magnitude of biomechanical plasticity per unit change of physical capacity will be higher during incline compared to level walking.

This chapter is separated into the following sections: old compared to young adults during level walking, old compared to young adults during incline walking, physical capacity scores, correlations during level walking at self-selected speeds (C1), correlations during level walking at a controlled speed – 1.30 m/s (C2), correlations during incline walking at self-selected speeds (C3), correlations during incline walking at a controlled speed – 1.20 m/s (C4), comparing level to incline walking at self-selected speeds, comparing level to incline walking at controlled speeds, and a summary.

Old compared to young adults during level walking

Student's t-tests were conducted to determine gait differences between young and old adults during level walking. These analyses reveal that, compared to young adults, old adults self-select a slower speed and walk with a shorter step length at both self-selected and controlled speeds. Because old adults took shorter steps during the controlled condition, they exhibited greater cadence compared to young adults in order to meet the controlled speed of 1.30 ms^{-1} (**Table 2**).

Table 2: Mean and standard deviation values of spatiotemporal variables in young versus old adults during level walking at self-selected speeds and a controlled speed (1.30 m/s). P-values are **bolded** to show significant differences ($p < 0.05$).

Spatiotemporal Variables During Level Walking										
Variables	Level - Self-Selected Speed					Level - Controlled Speed				
	Young	SD	Old	SD	P-Value	Young	SD	Old	SD	P-Value
Gait Velocity ($m \cdot s^{-1}$)	1.49	0.17	1.34	0.17	**0.01**	1.30	0.03	1.31	0.04	0.29
Step Length (m)	1.55	0.13	1.39	0.15	**0.00**	1.45	0.06	1.38	0.10	**0.02**
Cadence (steps/min)	115	5.00	116	9.00	0.47	108	3.00	114	8.00	**0.01**

Joint kinetic comparisons between young and old adults were conducted to ensure that age-associated biomechanical plasticity existed within this study’s old group during level walking (**Table 3**). While old adults walked at self-selected speeds, their hip extensors contributed 11.1% more ($p < 0.05$) of the total positive work during support while their ankle plantarflexors contributed 7.8% less ($p < 0.05$) of the total positive work during support compared to young adults (**Figures 2 and 3**). Also at self-selected speeds, old adults exhibited ~7% smaller peak plantarflexor toque ($p < 0.05$), ~25% smaller peak plantarflexor positive power ($p < 0.01$), and ~22% smaller plantarflexor positive work compared to young adults ($p < 0.01$). Finally, at self-selected speeds, old adults exhibited larger hip/ankle peak extensor positive power ($p < 0.05$) and extensor positive work ($p < 0.05$) ratios compared to young adults (**Figure 4**). These results indicate that age-associated biomechanical plasticity was present in this study’s old population during level walking at self-selected speeds.

While old adults walked at the controlled speed of 1.30 ms^{-1}, their hip extensors contributed 10.8% more ($p < 0.05$) of the total positive work during support while their ankle plantarflexors contributed 8.8% less ($p < 0.05$) of the total positive work during support compared to young adults. Also at the controlled speed, old adults exhibited ~35% larger peak hip extensor positive power ($p = 0.05$), ~46% more hip extensor positive work ($p = 0.05$), ~12% less plantarflexor angular impulse ($p < 0.05$), ~12% smaller peak plantarflexor positive power ($p < 0.05$), and ~9% less plantarflexor positive work ($p < 0.05$) compared to young adults. Finally, at the controlled speed, old adults exhibited a larger hip/ankle peak extensor torque ratio ($p = 0.05$), hip/ankle angular impulse ratio ($p < 0.05$), hip/ankle peak extensor positive power ratio ($p = 0.05$), and hip/ankle extensor positive work ratio ($p < 0.05$) compared to young adults. Average hip, knee, and ankle joint torque and power curves for old and young adults during level walking at self-selected and controlled speeds are displayed in **Figure 2**. These results indicate that age-associated biomechanical plasticity was present in this study's old population during level walking at the controlled speed.

Table 3: Mean and standard deviation values of joint kinetic variables in young versus old adults during level walking at self-selected speeds and a controlled speed (1.30 ms^{-1}). P-values are **bolded** to show significant differences ($p < 0.05$).

Comparisons of Joint Kinetic Variables During Level Walking											
Joint	Variables	Level - Self-Selected Speed					Level - Controlled Speed				
		Young	SD	Old	SD	P-Value	Young	SD	Old	SD	P-Value
Hip	Peak Extensor Torque (Nm · kg^{-1})	0.79	0.12	0.79	0.23	0.47	0.69	0.15	0.76	0.16	0.09
	Extensor Angular Impulse (Nm · s^{-1} · kg^{-1})	0.14	0.06	0.16	0.06	0.25	0.14	0.06	0.16	0.06	0.15
	Peak Extensor Positive Power (W · kg^{-1})	1.01	0.38	1.24	0.64	0.15	0.86	0.38	1.16	0.53	**0.05**
	Extensor Positive Work (J · kg^{-1})	0.14	0.09	0.19	0.10	0.10	0.13	0.09	0.19	0.10	**0.05**
	% of Total Extensor Positive Work	29.2	15.0	40.3	14.0	**0.02**	30.5	14.9	41.3	13.8	**0.02**
Ankle	Peak Plantarflexor Torque (Nm · kg^{-1})	1.56	0.19	1.45	0.16	**0.03**	1.49	0.13	1.43	0.13	0.14
	Plantarflexor Angular Impulse (Nm · s^{-1} · kg^{-1})	0.38	0.05	0.36	0.06	0.20	0.42	0.05	0.37	0.05	**0.01**
	Peak Plantarflexor Positive Power (W · kg^{-1})	3.50	0.90	2.64	0.44	**0.00**	2.88	0.48	2.53	0.38	**0.01**
	Plantarflexor Positive Work (J · kg^{-1})	0.27	0.06	0.21	0.04	**0.00**	0.22	0.03	0.20	0.03	**0.04**
	% of Total Extensor Positive Work	57.1	8.4	49.3	10.8	**0.02**	57.4	9.4	48.6	10.0	**0.01**
Hip/Ankle	Peak Torque Ratio	0.51	0.07	0.55	0.16	0.21	0.46	0.09	0.54	0.13	**0.05**
	Angular Impulse Ratio	0.38	0.13	0.45	0.19	0.13	0.33	0.15	0.44	0.17	**0.04**
	Peak Power Ratio	0.30	0.12	0.48	0.24	**0.01**	0.30	0.12	0.47	0.23	**0.02**
	Work Ratio	0.56	0.36	0.92	0.51	**0.02**	0.58	0.35	0.95	0.53	**0.02**

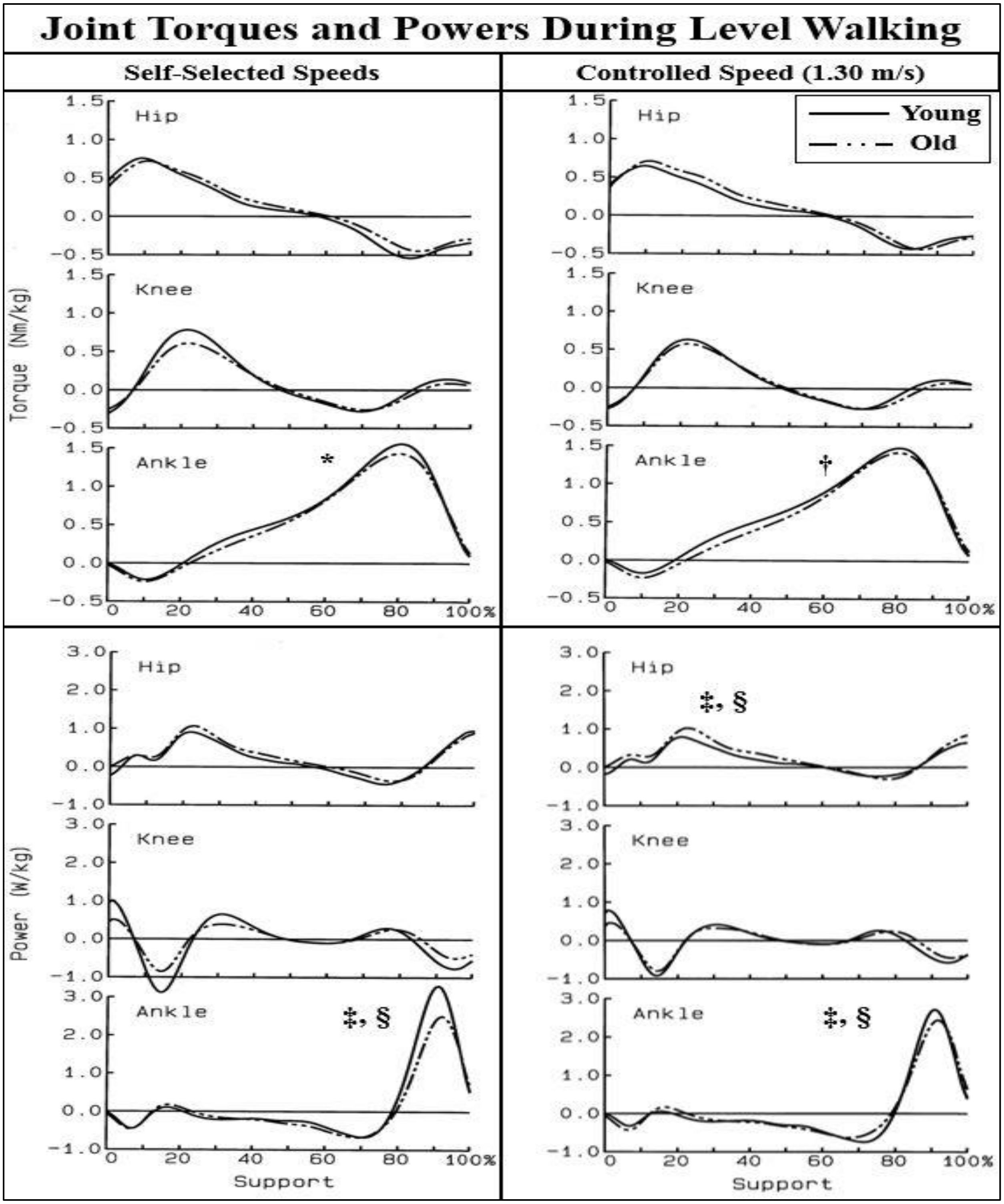

Figure 2: Hip, knee, and ankle joint torques (top row) and powers (bottom row) for level walking at self-selected (first column) and controlled (second column) speeds. Differences at the knee joint are not denoted. * indicates significant difference in peak torques; † indicates significant difference in angular impulse; ‡ indicates significant difference in peak powers; § indicates significant difference in work ($p < 0.05$).

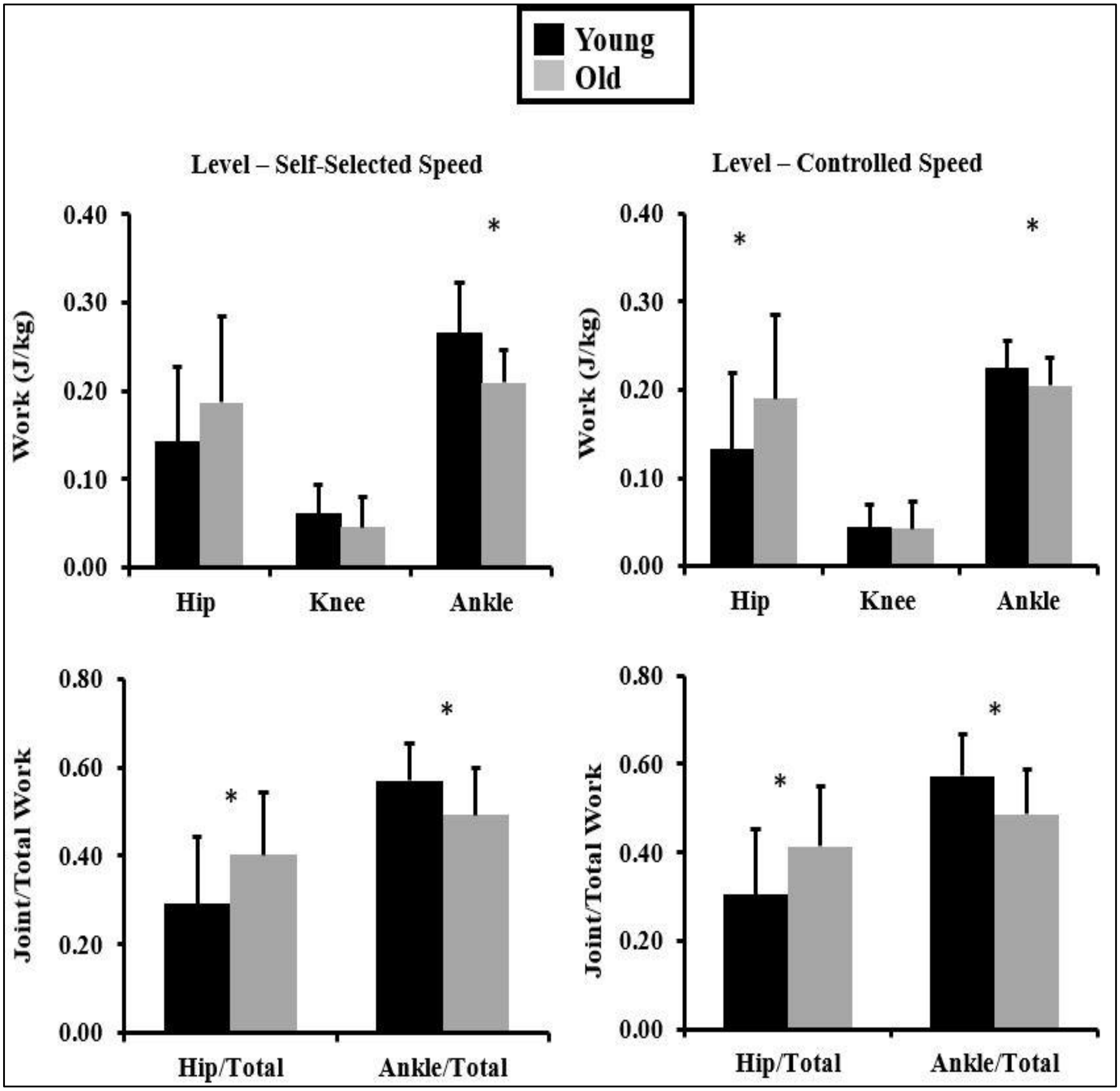

Figure 3: Hip, knee, ankle (top row), and hip/total, and ankle/total (bottom row) joint work values for young and old adults during level walking at self-selected (first column) and controlled (second column) speeds. Error bars represent +SD. * Indicates significant differences between young and old adults ($p < 0.05$).

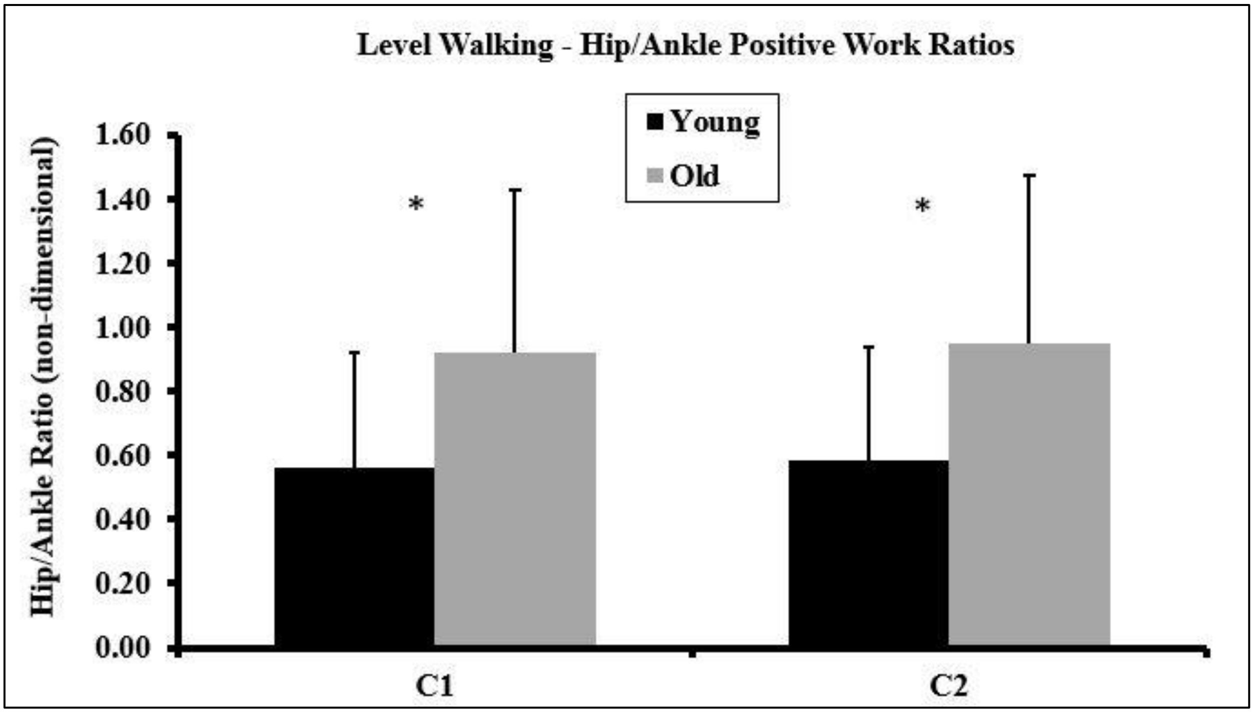

Figure 4: Hip/ankle positive extensor work ratios in the young group compared to the old group during level walking at self-selected (C1) and controlled (C2) speeds. Error bars represent +SD. * Indicates significant difference between young and old adults ($p < 0.05$).

Old compared to young adults during incline walking

Student's t-tests were also conducted to determine gait differences between old and young adults during the incline walking conditions. Similar to level walking, these analyses revealed that, compared to young adults, old adults self-selected slower speeds and walked with shorter step lengths at both self-selected and controlled speeds (**Table 4**).

Table 4: Mean and standard deviation values of spatiotemporal variables in young versus old adults during incline walking at self-selected speeds (C3) and a controlled speed (1.2 m/s; C4). P-values are **bolded** to show significant differences ($p < 0.05$).

Spatiotemporal Variables During Incline Walking										
Variables	Incline - Self-Selected Speed (C3)					Incline - Controlled Speed (C4)				
	Young	SD	Old	SD	P-Value	Young	SD	Old	SD	P-Value
Gait Velocity ($m \cdot s^{-1}$)	1.35	0.17	1.23	0.17	**0.02**	1.20	0.03	1.24	0.04	**0.01**
Step Length (m)	1.56	0.17	1.32	0.16	**0.00**	1.45	0.09	1.33	0.11	**0.00**
Cadence (steps/min)	103	5	111	11	**0.02**	100	8	112	9	**0.00**

Joint kinetic comparisons between old and young adults were conducted to ensure that age-associated biomechanical plasticity existed within this study's old group during incline walking (**Table 5**). While old adults walked up the incline at self-selected speeds, their hip extensors contributed 7.4% more ($p < 0.05$) of the total positive work during support while their ankle plantarflexors contributed 6.8% ($p < 0.01$) less of the total positive work during support compared to young adults (**Figure 6**). Also at self-selected speeds, old adults exhibited ~25% smaller peak plantarflexor positive power ($p < 0.01$) and ~25% less plantarflexor positive work ($p < 0.01$) compared to young adults. Finally, at self-selected speeds, old adults exhibited a larger hip/ankle peak extensor positive power ratio ($p < 0.05$) and a larger hip/ankle extensor positive work ratio ($p < 0.01$) compared to young adults (**Figure 7**). These results, particularly relative (to total) hip and ankle positive joint work and the two significantly different biomechanical plasticity ratios indicate that age-associated biomechanical plasticity existed in this study's old population during incline walking at self-selected speeds.

While old adults walked up the incline at the controlled speed of 1.20 ms^{-1}, their hip extensors contributed 9.5% more ($p < 0.01$) of the total positive work during support while their ankle plantarflexors contributed 7.5% less ($p < 0.01$) of the total positive work during support compared to young adults. Also while walking at the controlled speed, old adults exhibited ~15% larger peak hip extensor torque ($p < 0.05$), ~21% more hip extensor positive work ($p < 0.05$), ~14% less plantarflexor angular impulse ($p < 0.01$), ~20% smaller peak plantarflexor positive power ($p < 0.01$), and ~20% less plantarflexor positive work ($p < 0.01$) compared to young adults. Finally, at the controlled speed, old adults exhibited a larger hip/ankle peak extensor torque ratio ($p < 0.05$), a larger hip/ankle angular impulse ratio ($p < 0.05$), a larger hip/ankle peak extensor positive power ratio ($p < 0.01$), and a larger hip/ankle extensor positive work ratio compared to young adults ($p < 0.01$). Average hip, knee, and ankle joint torque and power curves for old and young adults during incline walking at self-selected and controlled speeds are displayed in **Figure 5**. These results indicate that age-associated biomechanical plasticity existed in this study's old population during incline walking at the controlled speed.

Table 5: Mean (SD) values of joint kinetic variables in young versus old adults during incline walking at self-selected speeds (C3) and a controlled speed (1.2 m/s; C4). P-values are **bolded** to show significant differences ($p < 0.05$).

Comparisons of Joint Kinetic Variables During Incline Walking											
		Incline - Self-Selected Speed					Incline - Controlled Speed				
Joint	Variables	Young	SD	Old	SD	P-Value	Young	SD	Old	SD	P-Value
Hip	Peak Extensor Torque ($Nm \cdot kg^{-1}$)	1.11	0.17	1.08	0.24	0.36	0.93	0.20	1.07	0.18	**0.02**
	Extensor Angular Impulse ($Nm \cdot s^{-1} \cdot kg^{-1}$)	0.27	0.08	0.28	0.07	0.33	0.25	0.08	0.28	0.07	0.14
	Peak Extensor Positive Power ($W \cdot kg^{-1}$)	2.46	0.60	2.41	0.84	0.43	2.08	0.57	2.39	0.62	0.09
	Extensor Positive Work ($J \cdot kg^{-1}$)	0.53	0.14	0.58	0.17	0.21	0.48	0.17	0.58	0.15	**0.04**
	% of Total Extensor Positive Work	38.4	6.4	45.8	7.7	**0.01**	36.6	7.9	46.1	8.6	**0.00**
Ankle	Peak Plantarflexor Torque ($Nm \cdot kg^{-1}$)	1.64	0.17	1.57	0.18	0.13	1.64	0.15	1.56	0.17	0.10
	Plantarflexor Angular Impulse ($Nm \cdot s^{-1} \cdot kg^{-1}$)	0.45	0.10	0.43	0.07	0.21	0.49	0.09	0.42	0.05	**0.00**
	Peak Plantarflexor Positive Power ($W \cdot kg^{-1}$)	4.29	1.10	3.22	0.62	**0.00**	4.09	0.75	3.26	0.62	**0.00**
	Plantarflexor Positive Work ($J \cdot kg^{-1}$)	0.57	0.12	0.43	0.09	**0.00**	0.54	0.09	0.43	0.10	**0.00**
	% of Total Extensor Positive Work	41.0	3.0	34.2	6.7	**0.00**	41.9	2.4	34.4	7.4	**0.00**
Hip/Ankle	Peak Torque Ratio	0.68	0.10	0.69	0.15	0.39	0.57	0.12	0.70	0.14	**0.01**
	Angular Impulse Ratio	0.61	0.18	0.66	0.20	0.22	0.51	0.18	0.67	0.20	**0.02**
	Peak Power Ratio	0.59	0.14	0.76	0.25	**0.02**	0.51	0.12	0.76	0.22	**0.00**
	Work Ratio	0.95	0.20	1.43	0.48	**0.00**	0.89	0.23	1.45	0.54	**0.00**

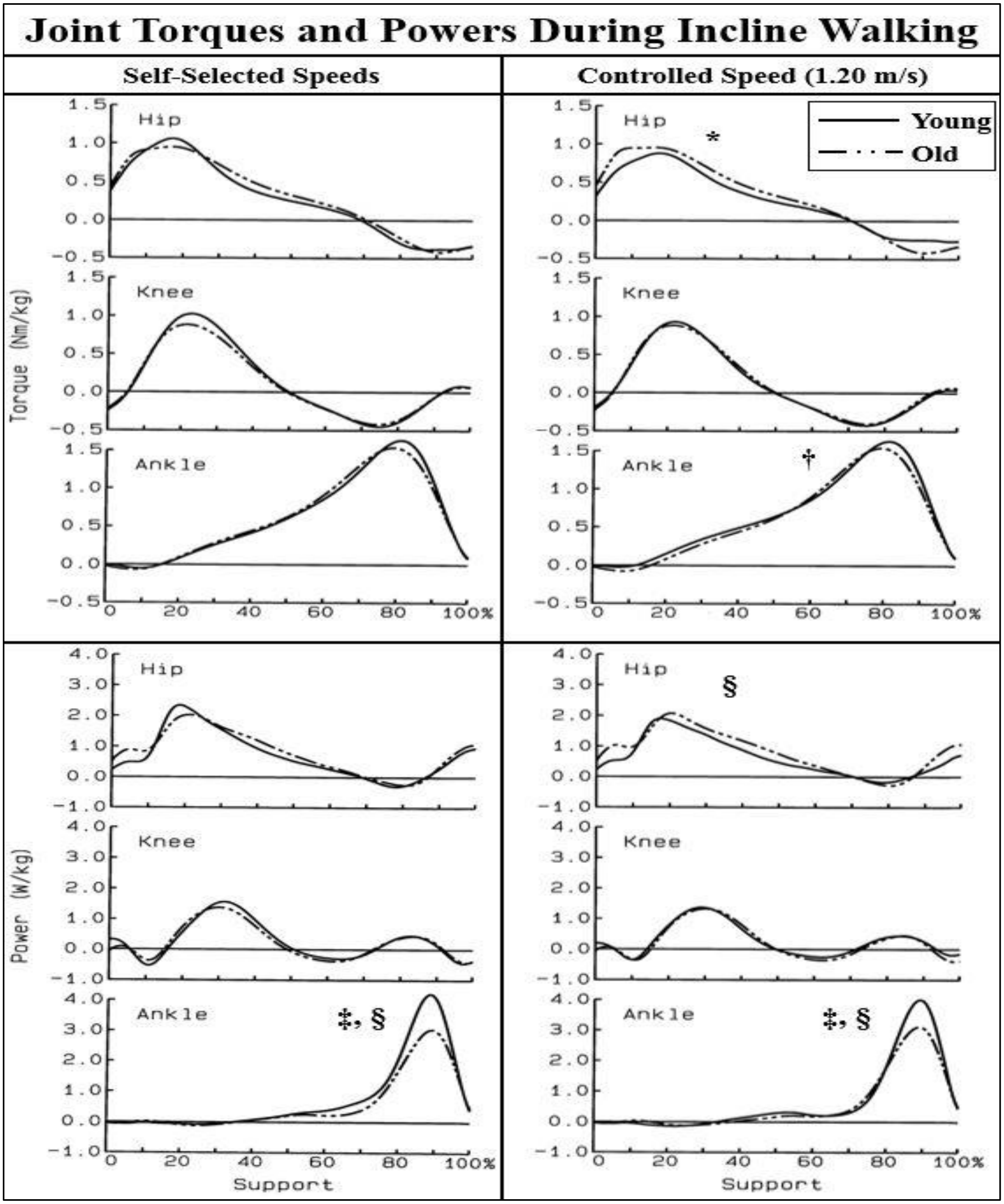

Figure 5: Hip, knee, and ankle joint torques (top row) and powers (bottom row) for incline walking at self-selected (first column) and controlled (second column) speeds. Differences at the knee joint are not denoted. * indicates significant difference in peak torques; † indicates significant difference in angular impulse; ‡ indicates significant difference in peak powers; § indicates significant difference in work ($p < 0.05$).

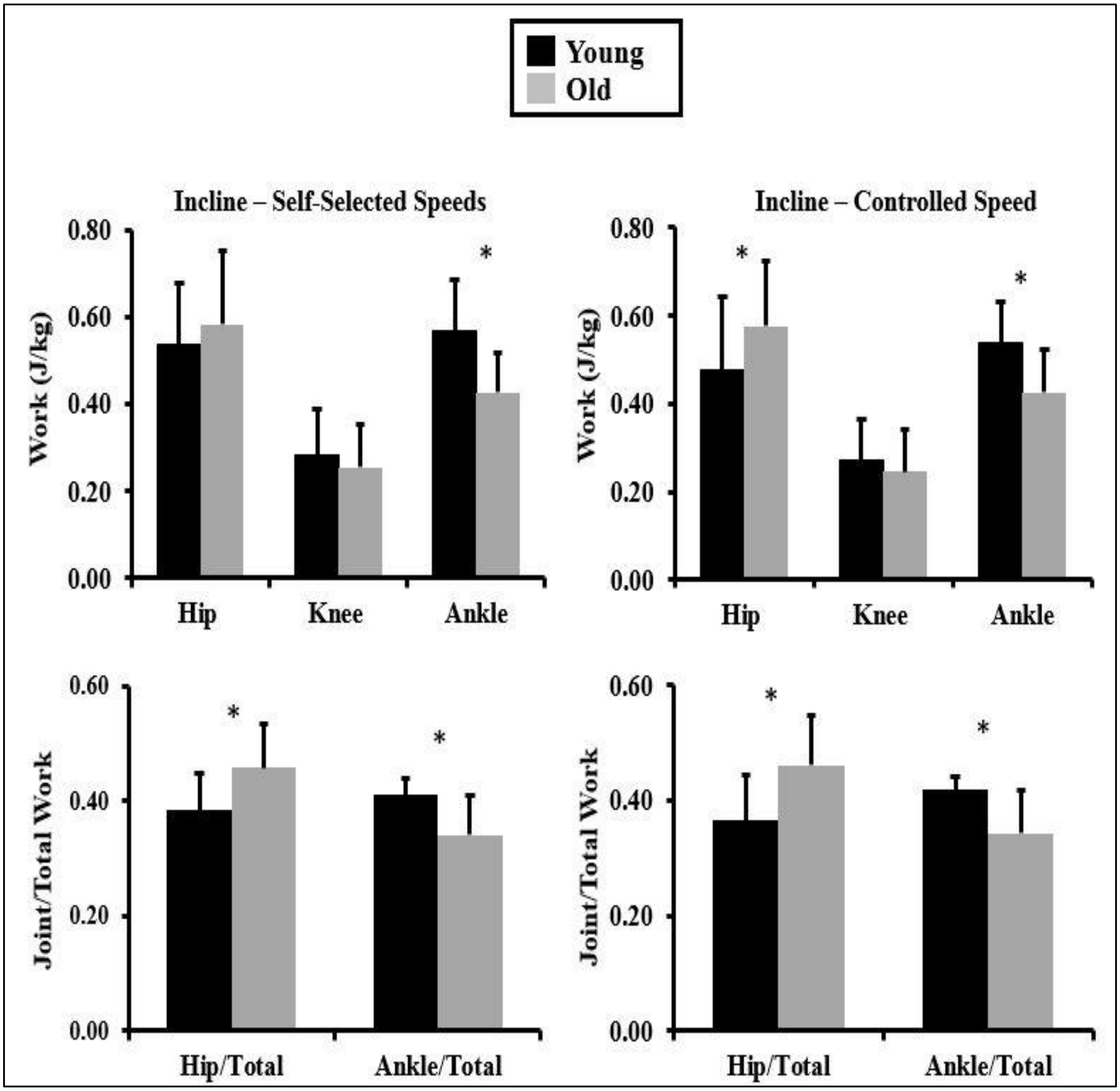

Figure 6: Hip, knee, ankle (top row), and hip/total and ankle/total (bottom row) joint work values for young and old adults during incline walking at self-selected (first column) and controlled (second column) speeds. Error bars represent +SD. * Indicates significant differences between young and old adults ($p < 0.05$).

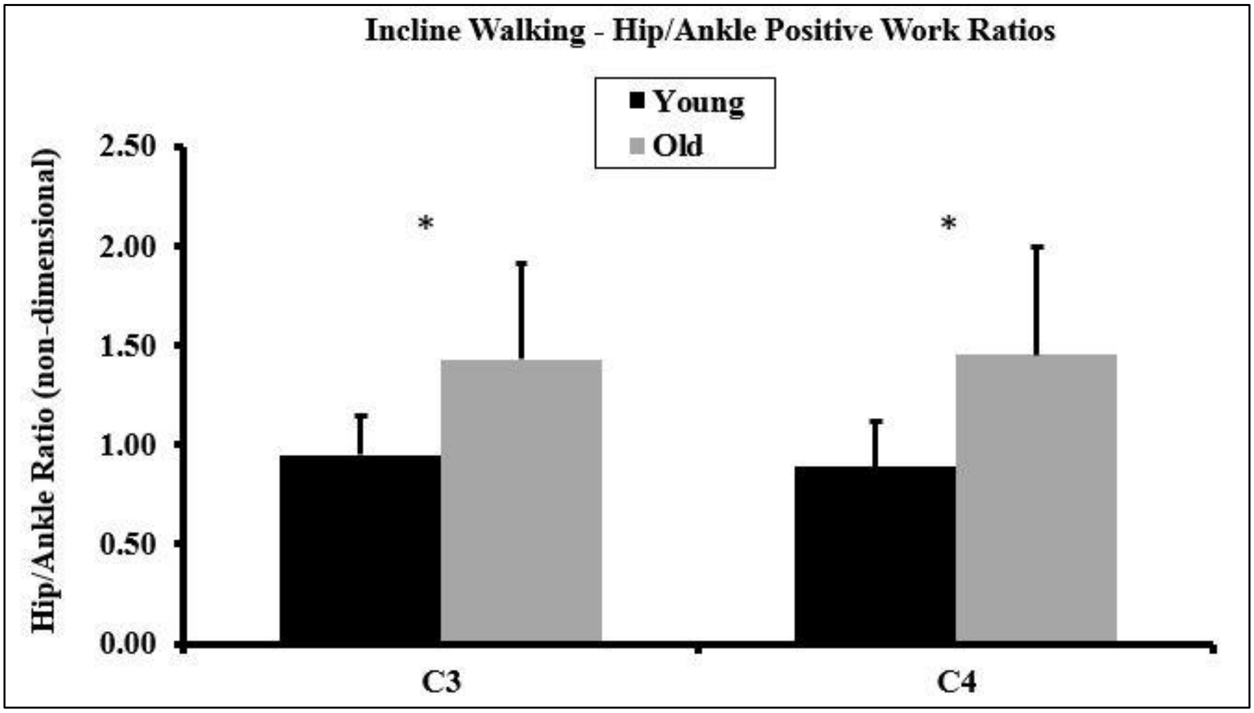

Figure 7: Hip/ankle joint positive work ratios in the young group compared to the old group during C3 and C4. Error bars represent +SD. * Indicates significant difference between young and old adults ($p < 0.05$).

Physical capacity scores

Short Form Health Survey physical component (SF-36 PC) scores and 20-meter self-selected walking speeds were chosen as measures of physical capacity within the old group. Visual representations of the spread in SF-36 PC scores (**Figure 8**) and 20-m self-selected speeds (**Figure 9**) are depicted below. These figures represent the fact that we included individuals across an adequate range of physical capacity and that there appears to be an even spread of physical capacity values across our sample of old adults (i.e. this sample does not appear to be clustered within a single area).

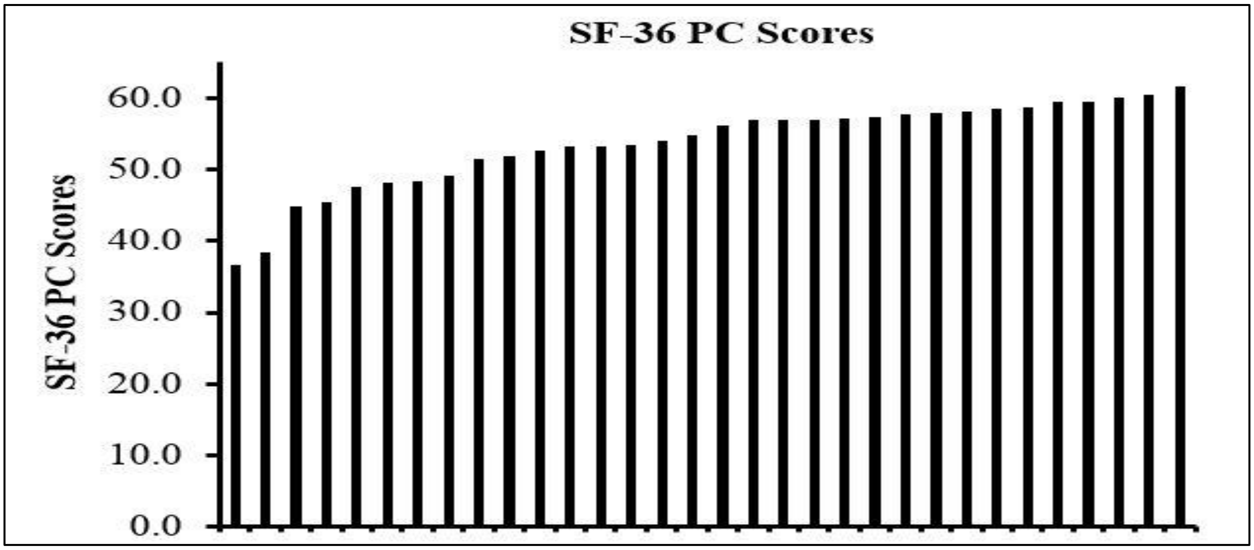

Figure 8: Short Form Health Survey physical component scores for all old participants.

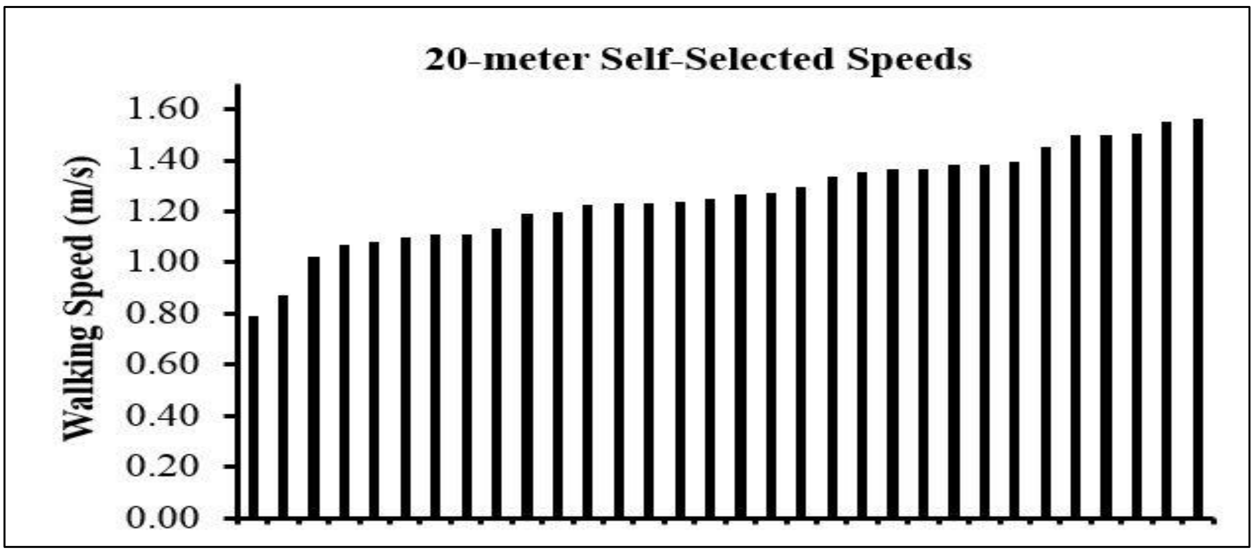

Figure 9: 20-meter self-selected walking speeds for all old participants.

Correlations during level walking at self-selected speeds

Within the old group, Pearson product moment correlations were calculated at 30 degrees of freedom (**Table 6**). These analyses were conducted to quantify relationships between physical capacity measures (SF-36 PC scores and 20-meter self-selected speeds) and the magnitude of biomechanical plasticity during level walking at self-selected speeds.

While using SF-36 PC scores as the explanatory variable in correlation analyses, significant positive relationships were found with the following joint kinetic variables: hip extensor peak torques, hip extensor angular impulses, hip extensor peak positive powers, hip extensor positive work, hip/total extensor positive work ratios, and ankle plantarflexor positive work. A significant negative relationship existed between SF-36 PC scores and ankle plantarflexor/total extensor positive work ratios. Scatter plots displaying the relationships between SF-36 scores and all joint kinetic variables of interest are presented in **Figure 10**. Additionally, significant positive relationships existed between SF-36 PC scores and the following biomechanical plasticity ratios: hip/ankle peak torque ratios, hip/ankle angular impulse ratios, hip/ankle peak extensor positive power ratios, and hip/ankle extensor positive work ratios (**Figure 12**).

While using 20-meter self-selected walking speeds as the explanatory variable in correlation analyses, significant positive relationships were found with the following joint kinetic variables: hip extensor peak torques, hip extensor peak positive powers, ankle plantarflexor peak torques, ankle plantarflexor peak positive powers, ankle plantarflexor positive work (**Table 6**). Significant negative relationships existed between 20-meter self-selected walking speeds and the following joint kinetic variables: ankle plantarflexor angular impulses and ankle/total extensor positive work ratios. Scatter plots displaying the relationships between 20-meter self-selected walking speeds and all joint kinetic variables of interest are presented in **Figure 11.** Additionally, significant positive relationships existed between 20-meter self-selected walking speeds and the following biomechanical plasticity ratios: hip/ankle peak torque ratios, hip/ankle angular impulse ratios, and hip/ankle peak extensor positive power ratios.

Table 6: Pearson's correlation coefficients (r) and r^2 values between physical capacity measures (SF-36 PC scores and 20-m self-selected walking speeds) and joint kinetic variables during level walking at self-selected speeds. R-values of statistically significant correlations are displayed in **bold** print ($p < 0.05$).

Correlation Analysis of Level Walking at Self-Selected Speeds					
		SF-36		20-m Self-Selected Speed	
Joint	Variable	r-value	r^2	r-value	r^2
Hip	Peak Extensor Torque (Nm·kg-1)	**0.47**	0.22	**0.55**	0.30
	Extensor Angular Impulse (Nm·s-1·kg-1)	**0.44**	0.19	0.22	0.05
	Peak Extensor Positive Power (W·kg-1)	**0.46**	0.21	**0.43**	0.19
	Extensor Positive Work (J·kg-1)	**0.42**	0.17	0.29	0.09
	% of Total Extensor Positive Work	**0.31**	0.10	0.17	0.03
Ankle	Peak Plantarflexor Torque (Nm·kg-1)	0.27	0.07	**0.34**	0.11
	Plantarflexor Angular Impulse (Nm·s-1·kg-1)	0.09	0.01	**-0.32**	0.10
	Peak Planterflexor Positive Power (W·kg-1)	0.08	0.01	**0.33**	0.11
	Plantarflexor Positive Work (J·kg-1)	**0.35**	0.12	**0.31**	0.10
	% of Total Extensor Positive Work	**-0.38**	0.15	**-0.31**	0.10
Hip/Ankle	Peak Torque Ratio	**0.39**	0.15	**0.46**	0.21
	Angular Impulse Ratio	**0.37**	0.14	**0.35**	0.13
	Peak Power Ratio	**0.46**	0.21	**0.36**	0.13
	Work Ratio	**0.31**	0.09	0.21	0.05

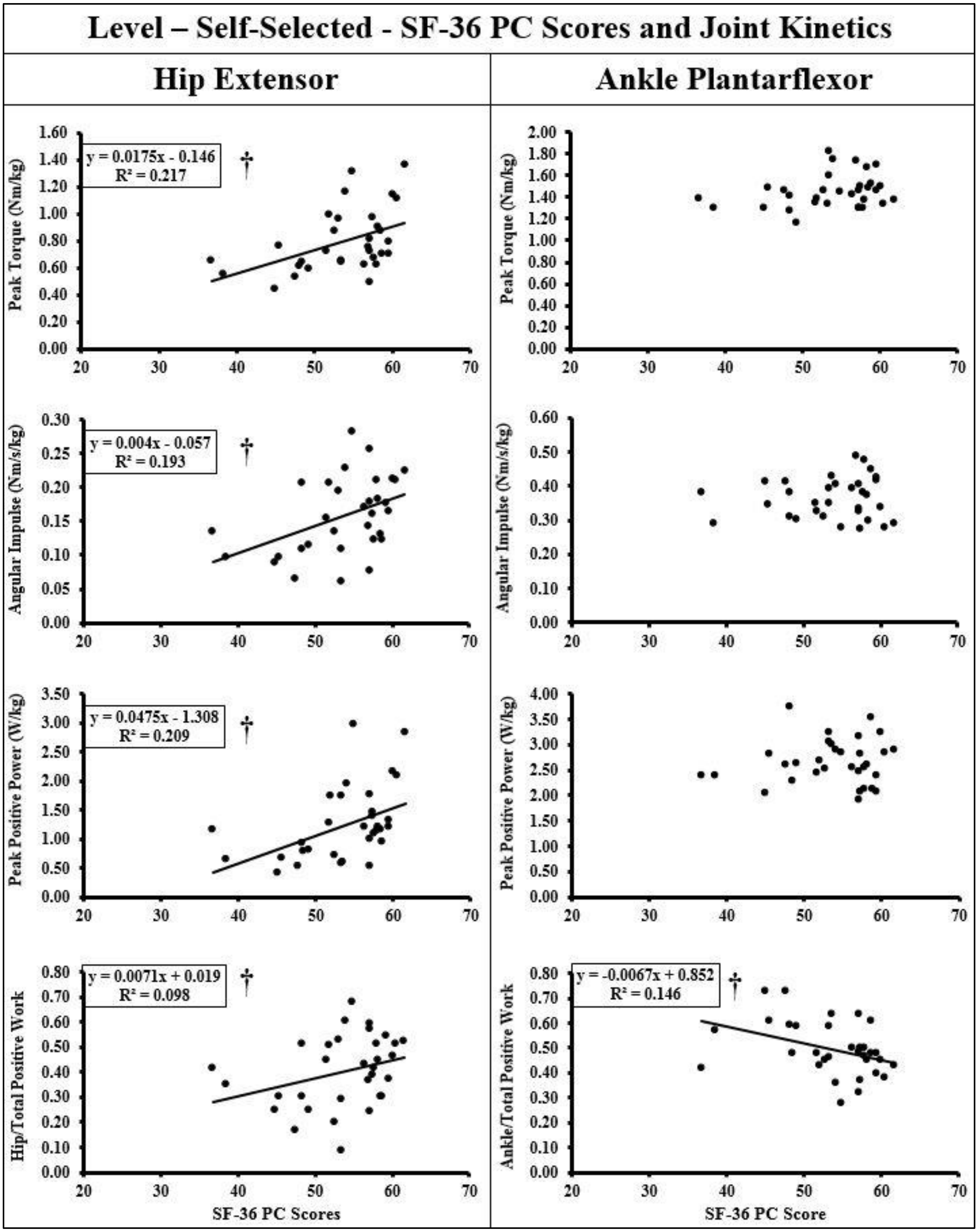

Figure 10: Scatter plots displaying correlations between SF-36 PC scores and hip (column one) and ankle (column two) peak torques, angular impulses, peak positive powers, and percent contribution to total extensor positive work during level walking at self-selected speeds. † Denotes a significant relationship ($p < 0.05$).

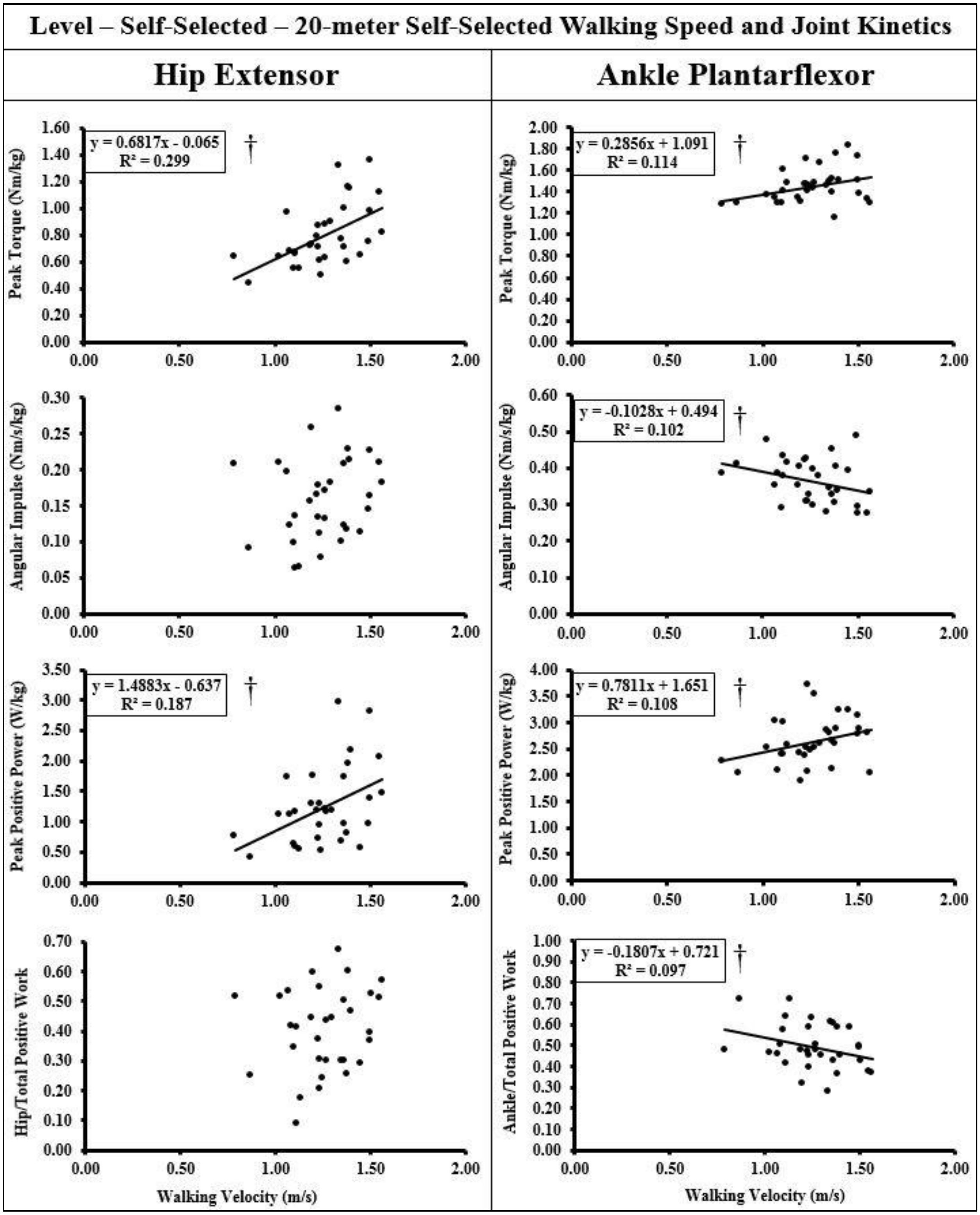

Figure 11: Scatter plots displaying correlations between 20-meter self-selected walking speeds and hip (column one) and ankle (column two) peak torques, angular impulses, peak positive powers, and percent contribution to total extensor positive work during level walking at self-selected speeds. † Denotes a significant relationship ($p < 0.05$).

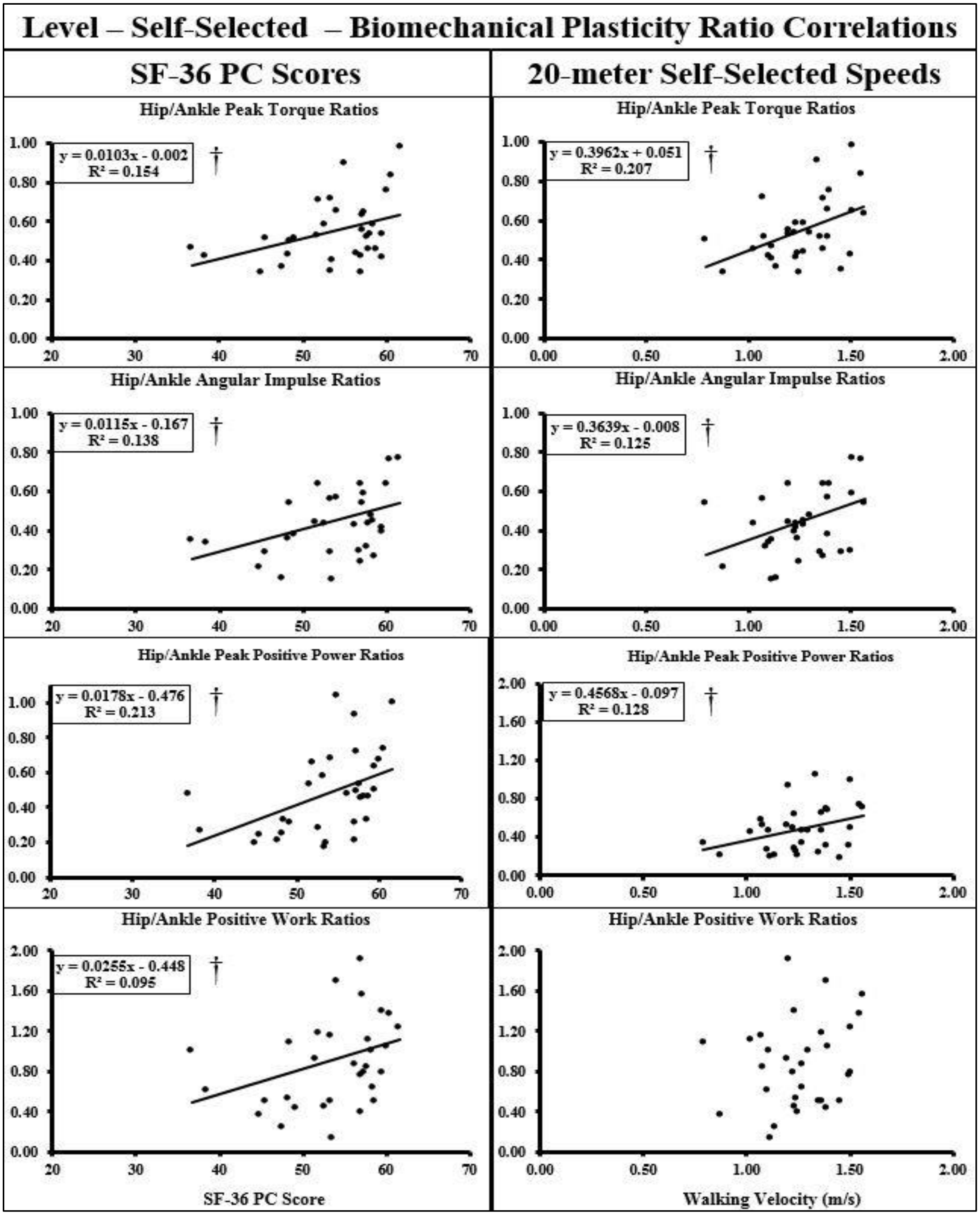

Figure 12: Scatter plots displaying correlations between measures of physical capacity (SF-36 PC scores – column one; 20-meter self-selected walking speed – column two) and biomechanical plasticity ratios during level walking at self-selected speeds. † Denotes a significant relationship ($p < 0.05$).

Correlations during level walking at a controlled speed – 1.30 m/s

Within the old group, Pearson product moment correlations were calculated at 30 degrees of freedom. These analyses were conducted to quantify the relationships between physical capacity measures (SF-36 PC scores and 20-meter self-selected speeds) and biomechanical plasticity during level walking at a controlled speed of 1.30 m/s (± 5%). All correlation coefficients for this condition are reported in **Table 7**.

While using SF-36 PC scores as the explanatory variable in correlation analyses, significant positive relationships were found with the following joint kinetic variables: hip extensor peak torques, hip extensor angular impulses, hip extensor peak positive powers, hip extensor positive work, ankle plantarflexor angular impulses, and ankle plantarflexor positive work. Scatter plots displaying the relationships between SF-36 scores and joint kinetic variables of interest are presented in **Figure 13.** Additionally, significant positive relationships existed between SF-36 PC scores the following biomechanical plasticity ratios: hip/ankle peak extensor positive power ratios (**Figure 15**).

While using 20-meter self-selected walking speeds as the explanatory variable in correlation analyses, no significant relationships existed with any of the joint kinetic variables of interest (**Figure 14**) or biomechanical plasticity ratios during level walking at the controlled speed.

Table 7: Pearson's correlation coefficients (r) and r^2 values between physical capacity measures (SF-36 PC scores and 20-m self-selected walking speeds) and joint kinetic variables during level walking at the controlled speed of 1.30 $m \cdot s^{-1}$ (± 5%). R-values of statistically significant correlations are displayed in **bold** print ($p < 0.05$).

Correlation Analysis of Level Walking at a Controlled Speed (1.30 m/s)					
		SF-36		20-m Self-Selected Speed	
Joint	Variable	r-value	r^2	r-Value	r^2
Hip	Peak Extensor Torque (Nm·kg-1)	**0.35**	0.13	0.16	0.03
	Extensor Angular Impulse (Nm·s-1·kg-1)	**0.41**	0.17	0.17	0.03
	Peak Extensor Positive Power (W·kg-1)	**0.41**	0.17	0.20	0.04
	Extensor Positive Work (J·kg-1)	**0.38**	0.15	0.17	0.03
	% of Total Extensor Positive Work	0.29	0.08	0.16	0.03
Ankle	Peak Plantarflexor Torque (Nm·kg-1)	0.21	0.04	0.15	0.02
	Plantarflexor Angular Impulse (Nm·s-1·kg-1)	**0.36**	0.13	0.09	0.01
	Peak Planterflexor Positive Power (W·kg-1)	-0.21	0.04	-0.22	0.05
	Plantarflexor Positive Work (J·kg-1)	**0.32**	0.10	0.02	0.00
	% of Total Extensor Positive Work	-0.29	0.08	-0.16	0.03
Hip/Ankle	Peak Torque Ratio	0.25	0.06	0.10	0.01
	Angular Impulse Ratio	0.27	0.07	0.15	0.02
	Peak Power Ratio	**0.43**	0.18	0.27	0.07
	Work Ratio	0.27	0.07	0.19	0.04

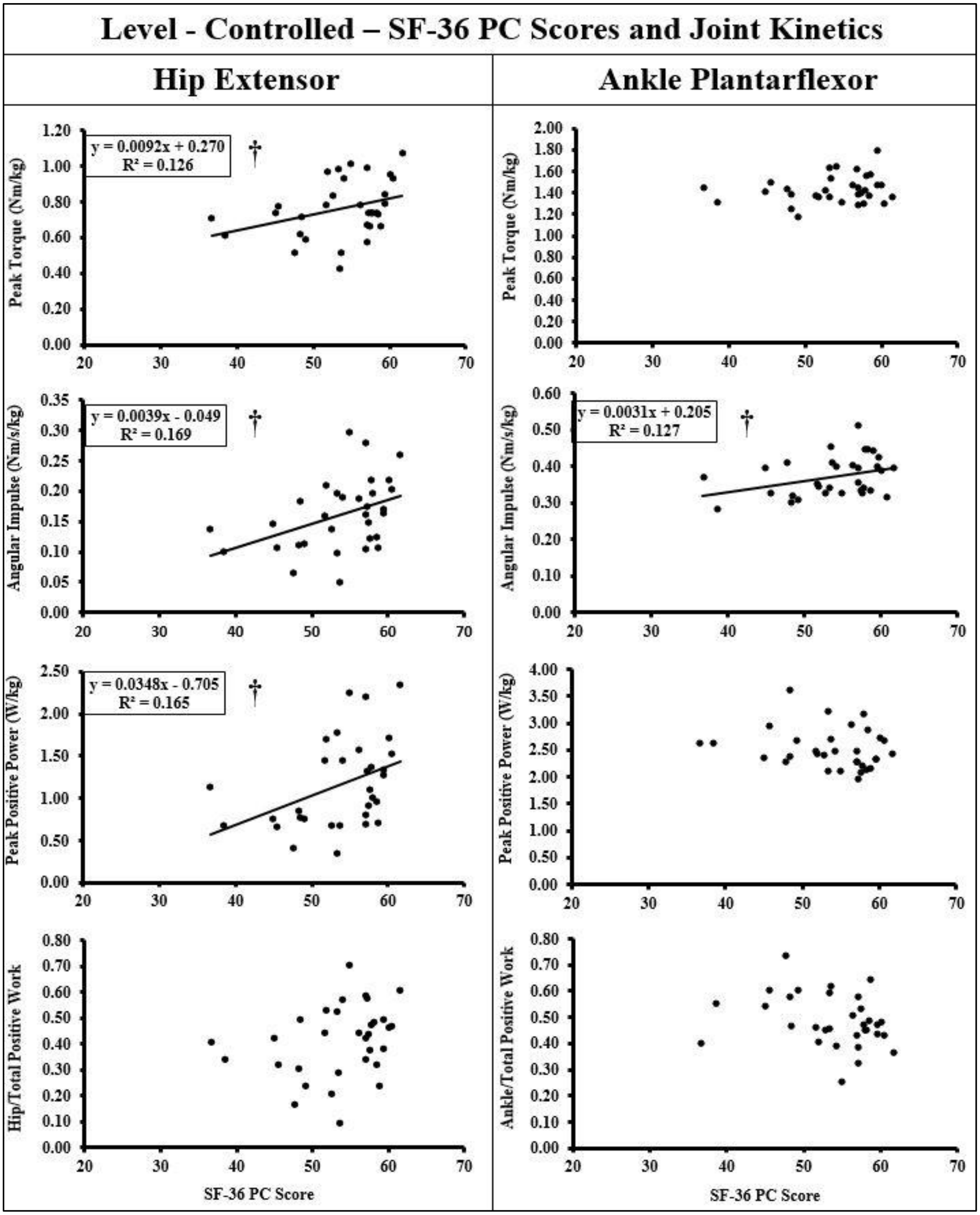

Figure 13: Scatter plots displaying correlations between SF-36 PC scores and hip (column one) and ankle (column two) peak torques, angular impulses, peak positive powers, and percent contribution to total extensor positive work during level walking at a controlled speed. † Denotes a significant relationship ($p < 0.05$).

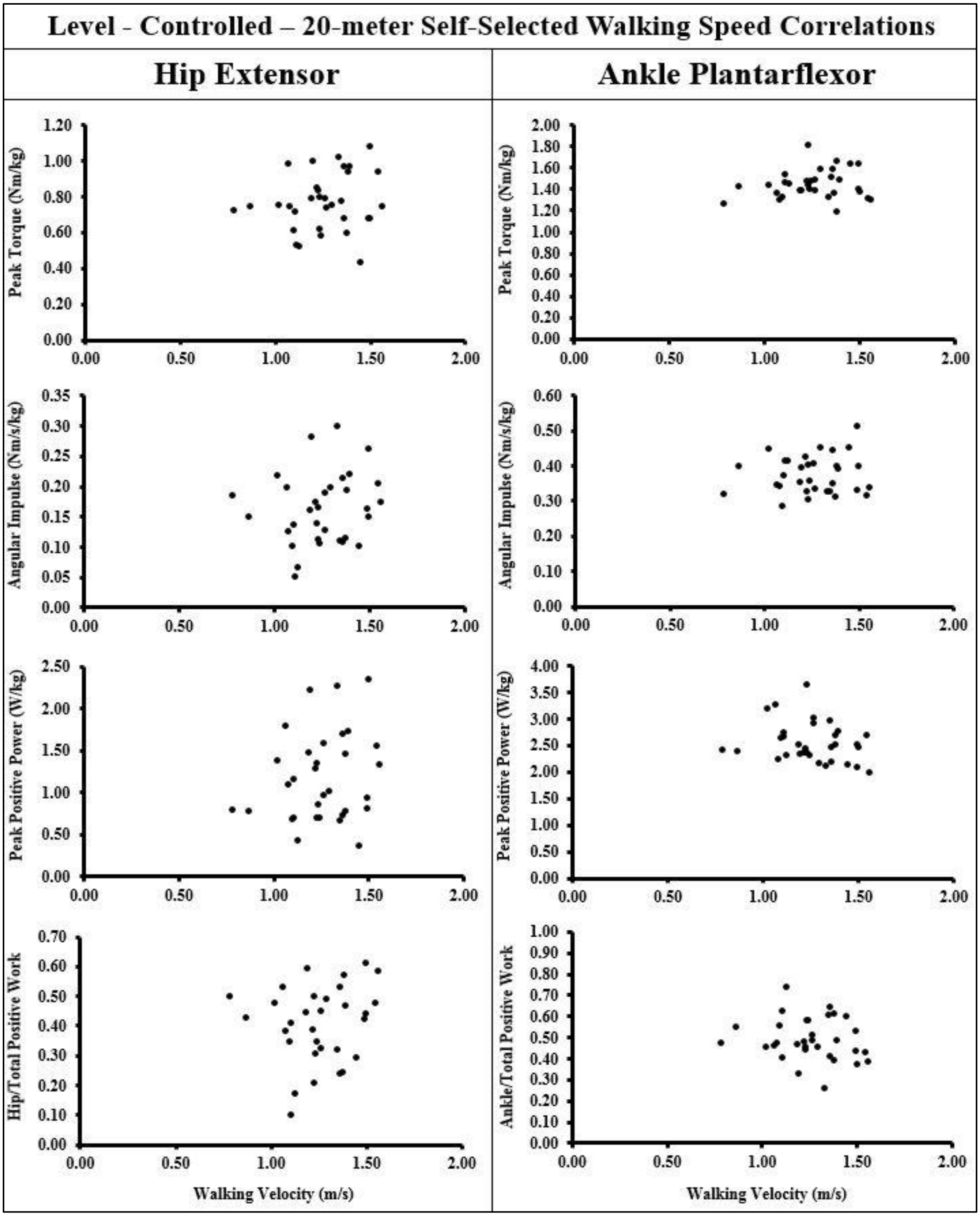

Figure 14: Scatter plots displaying correlations between 20-meter self-selected walking speeds and hip (column one) and ankle (column two) peak torques, angular impulses, peak positive powers, and percent contribution to total extensor positive work during level walking at a controlled speed.

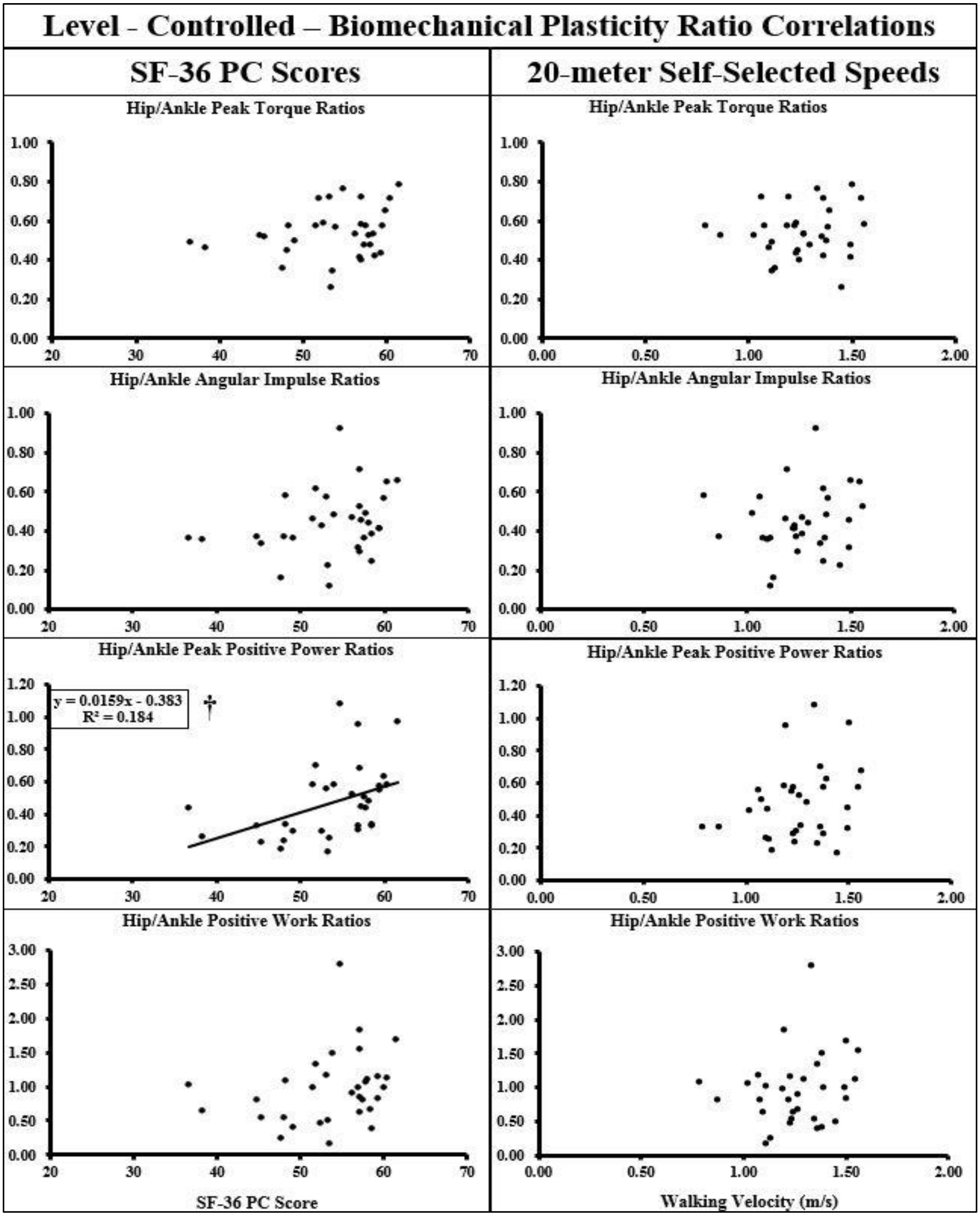

Figure 15: Scatter plots displaying correlations between measures of physical capacity (SF-36 PC scores – column one; 20-meter self-selected walking speed – column two) and biomechanical plasticity ratios during level walking at a controlled speed. † Denotes a significant relationship ($p < 0.05$).

Correlations during incline walking at self-selected speeds

Within the old group, Pearson product moment correlations were calculated at 29 degrees of freedom. These analyses were computed at 29 degrees of freedom due to the loss of data due to technical difficulty for one of our old participants. These analyses were conducted to quantify the relationships between physical capacity measures (SF-36 PC scores and 20-meter self-selected speeds) and biomechanical plasticity during incline walking at self-selected speeds. All correlation coefficients for this condition are reported in **Table 8**.

While using SF-36 PC scores as the explanatory variable in correlation analyses, significant positive relationships were found with the following joint kinetic variables: hip extensor peak torques, hip extensor angular impulses, hip extensor peak positive powers, hip extensor positive work, and ankle plantarflexor positive work. Scatter plots displaying the relationships between SF-36 PC scores and joint kinetic variables of interest are presented in **Figure 16.** Additionally, significant positive relationships existed between SF-36 PC scores and the following biomechanical plasticity ratios: hip/ankle peak torque ratios, hip/ankle angular impulse ratios, and hip/ankle peak positive power ratios (**Figure 18**).

While using 20-meter self-selected walking speeds as the explanatory variable in correlation analyses, significant positive relationships existed for the following joint kinetic variables: hip extensor peak torques, hip extensor peak positive powers, and ankle plantarflexor positive work. Scatter plots displaying the relationships between SF-36 PC scores and joint kinetic variables of interest are presented in **Figure 17.** No significant relationships existed between 20-meter self-selected speeds and biomechanical plasticity ratios during this condition.

Table 8: Pearson's correlation coefficients (r) and r^2 values between physical capacity measures (SF-36 PC scores and 20-m self-selected walking speeds) and joint kinetic variables during incline walking at self-selected speeds. R-values of statistically significant correlations are displayed in **bold** print ($p < 0.05$).

Correlation Analysis of Incline Walking at Self-Selected Speeds					
		SF-36		20-m Self-Selected Speed	
Joint	**Variable**	**r-value**	**r^2**	**r-value**	**r^2**
Hip	**Peak Extensor Torque (Nm·kg-1)**	**0.55**	0.31	**0.36**	0.13
	Extensor Angular Impulse (Nm·s-1·kg-1)	**0.36**	0.13	-0.02	0.00
	Peak Extensor Positive Power (W·kg-1)	**0.41**	0.17	**0.36**	0.13
	Extensor Positive Work (J·kg-1)	**0.49**	0.24	0.28	0.08
	% of Total Extensor Positive Work	0.28	0.08	-0.08	0.01
Ankle	**Peak Plantarflexor Torque (Nm·kg-1)**	0.16	0.02	0.18	0.03
	Plantarflexor Angular Impulse (Nm·s-1·kg-1)	-0.01	0.00	-0.07	0.00
	Peak Planterflexor Positive Power (W·kg-1)	0.23	0.05	0.20	0.04
	Plantarflexor Positive Work (J·kg-1)	**0.44**	0.19	**0.42**	0.18
	% of Total Extensor Positive Work	-0.02	0.00	0.06	0.00
Ratios	**Peak Torque Ratio**	**0.51**	0.26	0.30	0.09
	Angular Impulse Ratio	**0.32**	0.10	0.03	0.00
	Peak Power Ratio	**0.34**	0.12	0.29	0.08
	Work Ratio	0.15	0.02	-0.06	0.00

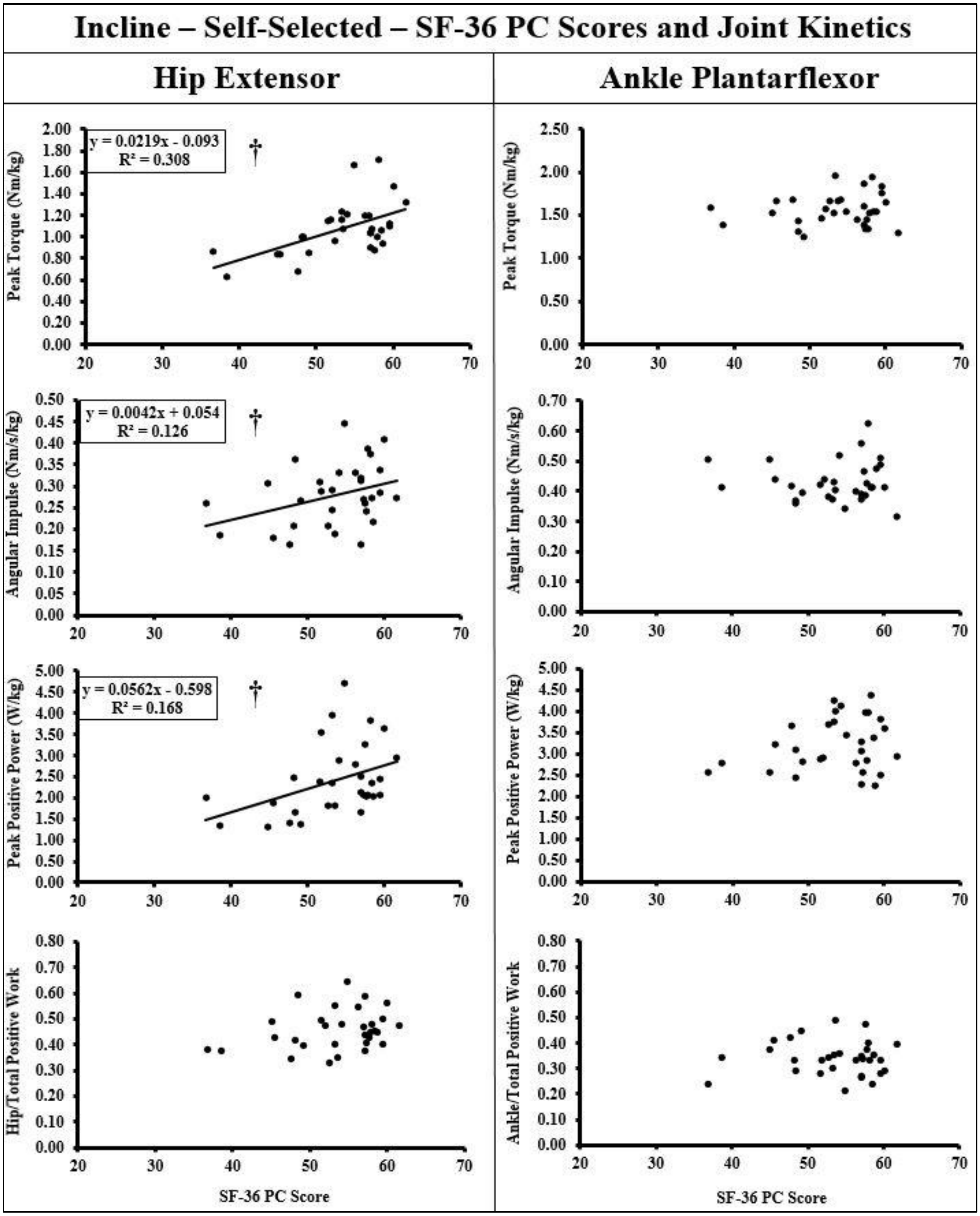

Figure 16: Scatter plots displaying correlations between SF-36 PC scores and hip (column one) and ankle (column two) peak torques, angular impulses, peak positive powers, and percent contribution to total extensor positive work during incline walking at self-selected speeds. † Denotes a significant relationship ($p < 0.05$).

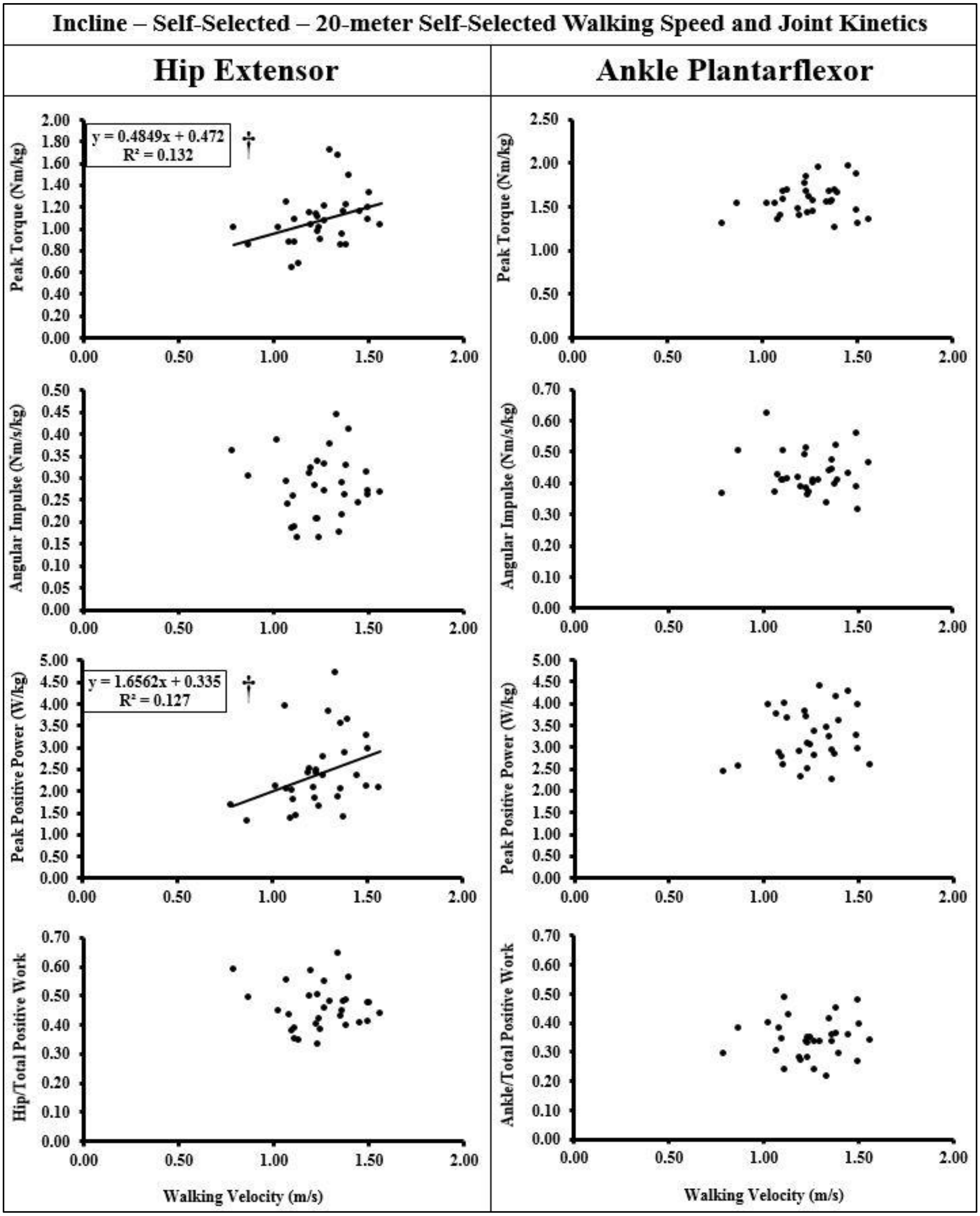

Figure 17: Scatter plots displaying correlations between 20-meter self-selected walking speeds and hip (column one) and ankle (column two) peak torques, angular impulses, peak positive powers, and percent contribution to total extensor positive work during incline walking at self-selected speeds. † Denotes a significant relationship ($p < 0.05$).

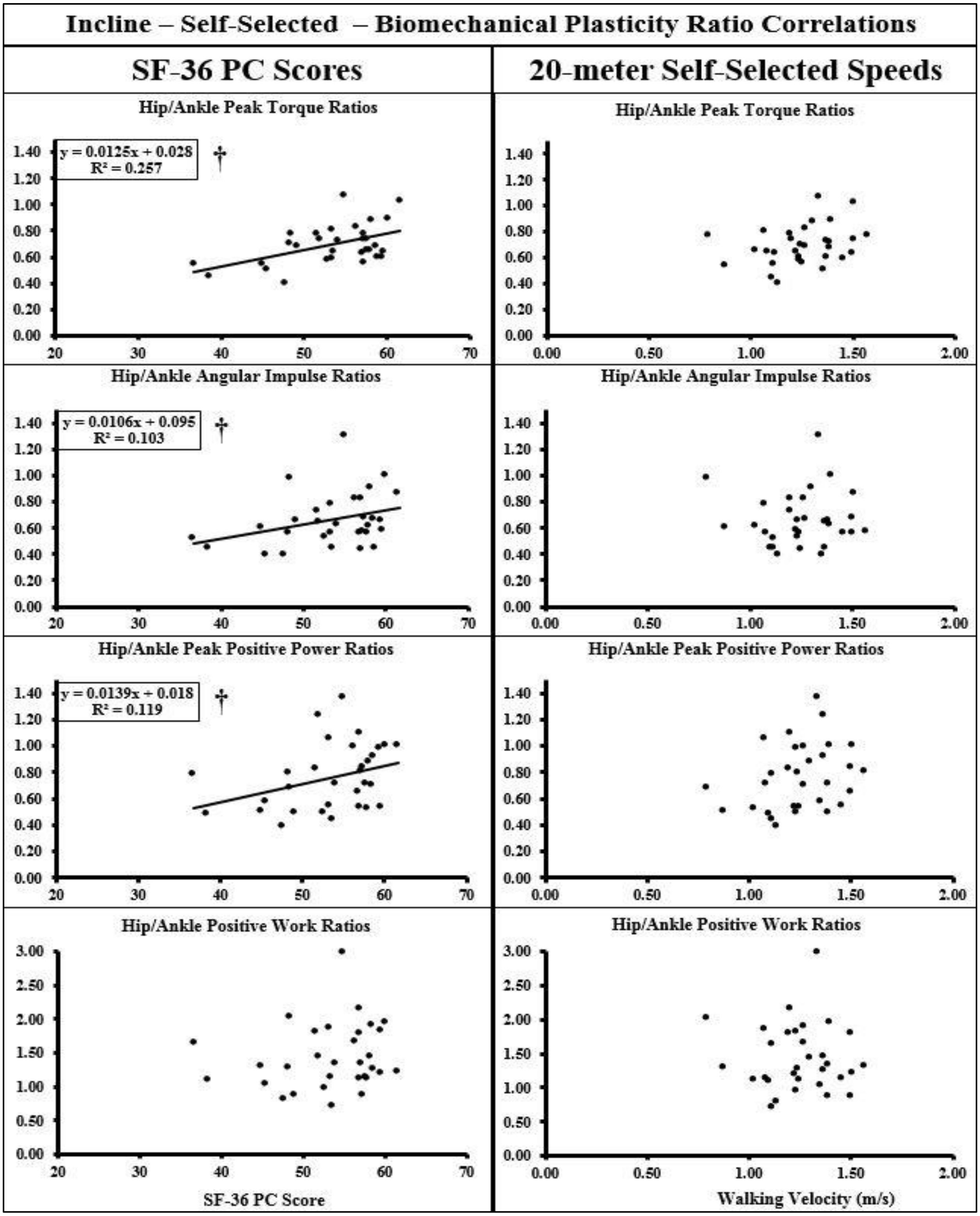

Figure 18: Scatter plots displaying correlations between measures of physical capacity (SF-36 PC scores – column one; 20-meter self-selected walking speed – column two) and biomechanical plasticity ratios during incline walking at self-selected speeds. † Denotes a significant relationship ($p < 0.05$).

Correlations during incline walking at a controlled speed – 1.20 m/s

Within the old group, Pearson product moment correlations were calculated at 29 degrees of freedom. These analyses were computed at 29 degrees of freedom due to the loss of data due to technical difficulty for one of our old participants. These analyses were conducted to quantify the relationships between physical capacity measures (SF-36 PC scores and 20-meter self-selected speeds) and biomechanical plasticity during incline walking at a controlled speed of 1.20 m/s (± 5%). All correlation coefficients for this condition are reported in **Table 9**.

While using SF-36 PC scores as the explanatory variable in correlation analyses, significant positive relationships existed with the following joint kinetic variables: hip extensor peak torques and ankle plantarflexor positive work. Scatter plots displaying the relationships between SF-36 PC scores and joint kinetic variables of interest are presented in **Figure 19.** No significant relationships existed between SF-36 PC scores and any of the biomechanical plasticity ratios (**Figure 21**).

While using 20-meter self-selected walking speeds as the explanatory variable in correlation analyses, no significant relationships existed with any of the joint kinetic variables of interest (**Figure 20**) or the biomechanical plasticity ratios.

Table 9: Pearson's correlation coefficients (r) and r^2 values between physical capacity measures (SF-36 PC scores and 20-m self-selected walking speeds) and joint kinetic variables during incline walking at the controlled speed of 1.20 $m \cdot s^{-1}$ (± 5%). R-values of statistically significant correlations are displayed in **bold** print ($p < 0.05$).

Correlation Analysis of Incline Walking at a Controlled Speed (1.20 m/s)					
		SF-36		20-m Self-Selected Speed	
Joint	Variable	r-value	r^2	r-value	r^2
Hip	Peak Extensor Torque (Nm·kg-1)	**0.37**	0.14	0.17	0.03
	Extensor Angular Impulse (Nm·s-1·kg-1)	0.29	0.08	-0.06	0.00
	Peak Extensor Positive Power (W·kg-1)	0.21	0.04	0.16	0.02
	Extensor Positive Work (J·kg-1)	0.30	0.09	0.07	0.01
	% of Total Extensor Positive Work	0.12	0.01	-0.19	0.04
Ankle	Peak Plantarflexor Torque (Nm·kg-1)	0.01	0.00	0.03	0.00
	Plantarflexor Angular Impulse (Nm·s-1·kg-1)	0.19	0.04	0.14	0.02
	Peak Planterflexor Positive Power (W·kg-1)	0.04	0.00	-0.03	0.00
	Plantarflexor Positive Work (J·kg-1)	**0.34**	0.12	0.22	0.05
	% of Total Extensor Positive Work	0.11	0.01	0.07	0.00
Hip/Ankle	Peak Torque Ratio	0.30	0.09	0.13	0.02
	Angular Impulse Ratio	0.17	0.03	-0.13	0.02
	Peak Power Ratio	0.17	0.03	0.13	0.02
	Work Ratio	0.00	0.00	-0.17	0.03

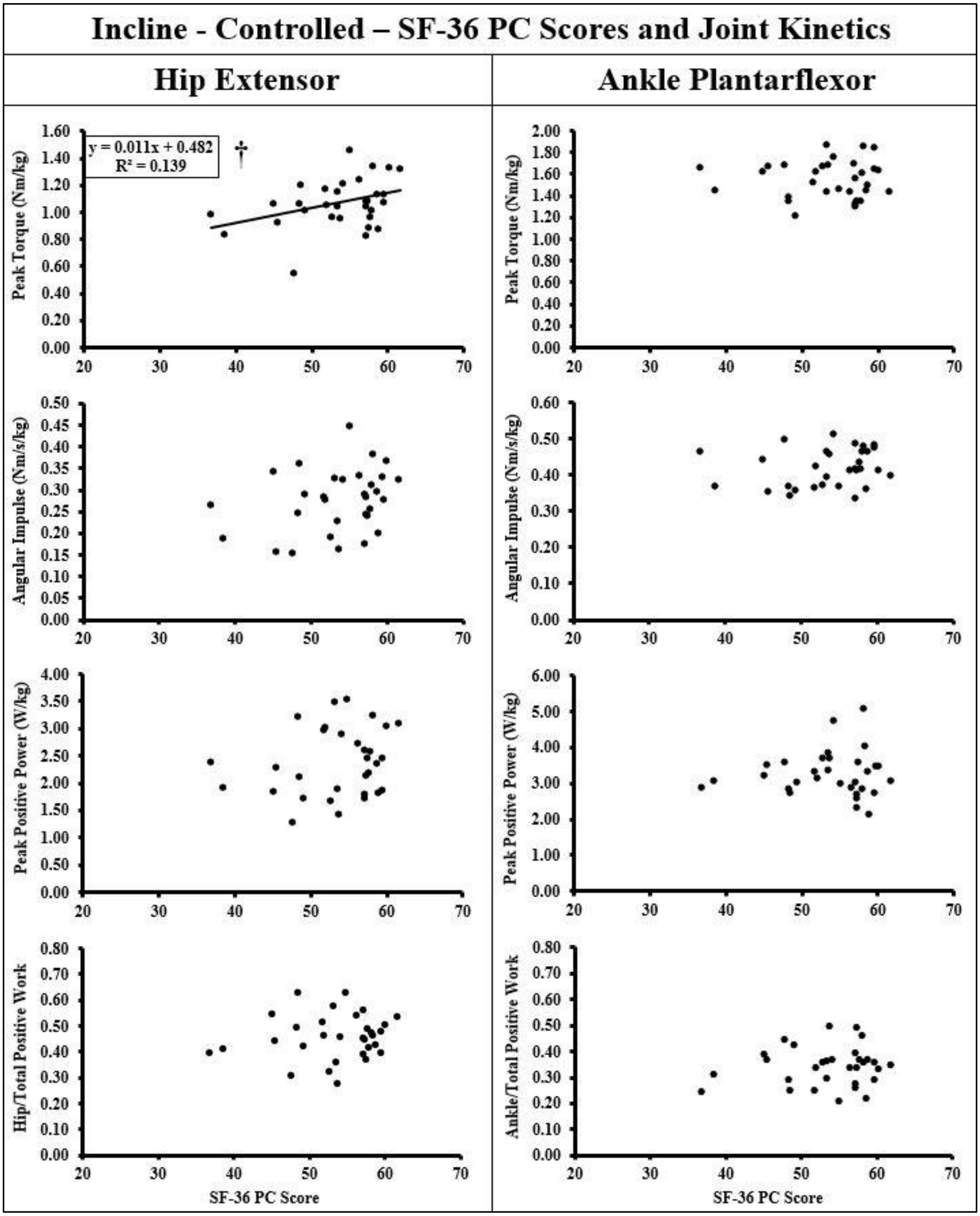

Figure 19: Scatter plots displaying correlations between SF-36 PC scores and hip (column one) and ankle (column two) peak torques, angular impulses, peak positive powers, and percent contribution to total extensor positive work during incline walking at a controlled speed. † Denotes a significant relationship ($p < 0.05$).

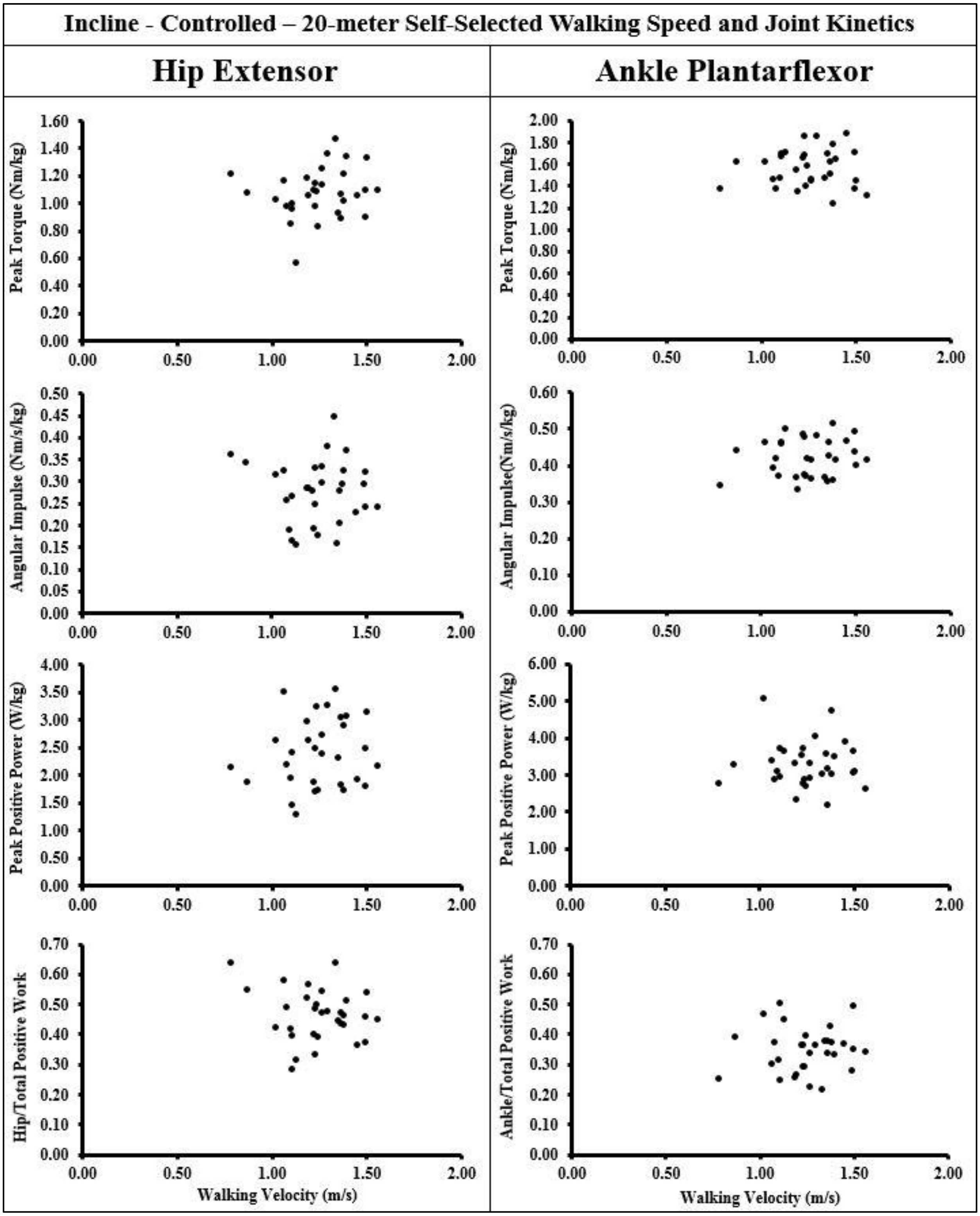

Figure 20: Scatter plots displaying correlations between 20-meter self-selected walking speeds and hip (column one) and ankle (column two) peak torques, angular impulses, peak positive powers, and percent contribution to total extensor positive work during incline walking at a controlled speed.

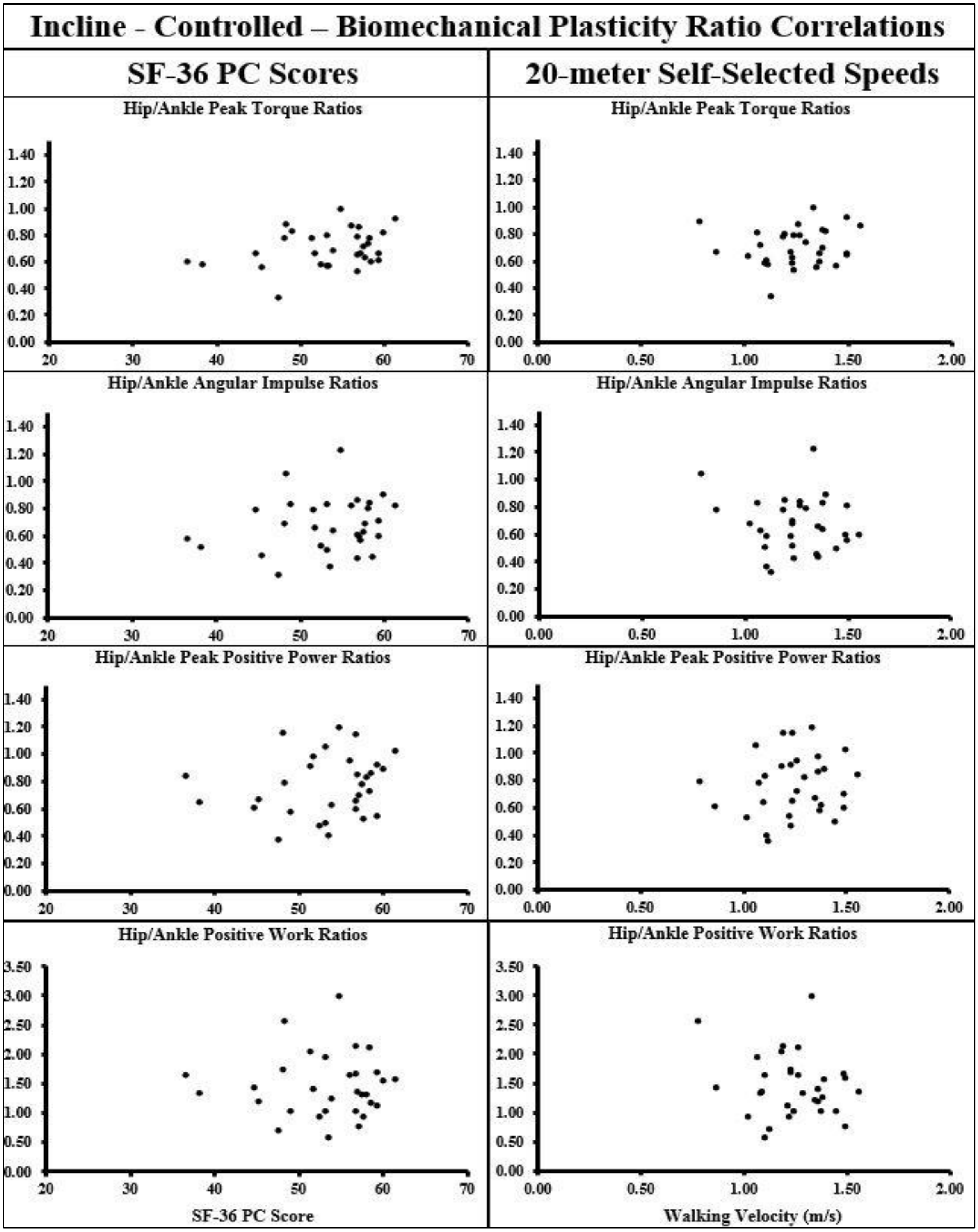

Figure 21: Scatter plots displaying correlations between measures of physical capacity (SF-36 PC scores – column one; 20-meter self-selected walking speed – column two) and biomechanical plasticity ratios during incline walking at a controlled speed.

Comparing level to incline walking at self-selected speeds

To test our second hypothesis that biomechanical plasticity becomes more pronounced during the more difficult task of incline compared to level walking, we compared the correlation results above between level and incline walking at self-selected speeds. Specifically, where corresponding significant relationships existed between level and incline conditions, we compared the 95% confidence intervals (C.I.s) of the slopes of the regressions (beta weights).

The following biomechanical plasticity ratios were significantly correlated to physical capacity during level *but not* incline walking at self-selected speeds: SF-36 PC scores vs. hip/ankle positive work ratio, 20-meter self-selected walking speed vs. hip/ankle peak torque ratio, 20-meter self-selected walking speed vs. hip/ankle angular impulse ratio, 20-meter self-selected walking speed vs. hip/ankle peak positive power ratio. Based on these four correlations, it seems possible that the relationship between physical capacity and biomechanical plasticity is stronger (or at least more observable) during level compared to incline walking. However, because three of the same correlation analyses revealed significant positive relationships during level *and* incline walking at self-selected speeds, we compared 95% C.I.s of the beta weights of these three correlations to determine if per unit changes in physical capacity resulted in larger per unit changes in biomechanical plasticity compared to level walking. Comparisons of the 95% C.I.s of regression slopes in these three correlations reveal large amounts of overlap (**Table 10**). Because of this, we conclude that there are no statistically significant differences between these correlations. Based on these findings, we cannot say conclusively that incline walking at self-selected speeds is associated an increased magnitude of biomechanical plasticity compared to level walking at self-selected speeds.

Table 10: Regression slopes with upper, median, and lower limits based on 95% C.I.s.

95% Confidence Intervals for Regression Slopes During Level and Incline Walking at Self-Selected Speeds						
	Level Walking at Self-Selected Speeds			Incline Walking at Self-Selected Speeds		
Variables	**Lower Limit**	**Median**	**Upper Limit**	**Lower Limit**	**Median**	**Upper Limit**
Hip/Ankle Peak Torque Ratios	0.0013	0.0103	0.0193	0.0044	0.0125	0.0205
Hip/Ankle Angular Impulse Ratios	0.0008	0.0115	0.0222	-0.0013	0.0106	0.0226
Hip/Ankle Peak Power Ratios	0.005	0.0178	0.0305	-0.0005	0.0139	0.0283

Comparing level to incline walking at controlled speeds

To test our second hypothesis that biomechanical plasticity becomes more pronounced during the more difficult task of incline compared to level walking, we compared the correlation results above between level and incline walking at controlled speeds. Specifically, we compared the 95% confidence intervals (C.I.s) of the slopes of the regressions (beta weights) computed for these conditions.

The following biomechanical plasticity ratios were significantly correlated to physical capacity during level but not incline walking at controlled speeds: SF-36 PC scores vs. hip/ankle peak positive power ratio. 95% C.I.s of the beta weights were not computed to compare level and incline walking at controlled speeds because there were no significant correlations corresponding between the two conditions. Based on these findings, we cannot say conclusively that incline walking at 1.20 m/s is associated an increased magnitude of biomechanical plasticity compared to level walking at 1.30 m/s.

Summary

Compared to young adults, old adults performed level and incline locomotion with an overall larger mechanical output at the hip and a lower mechanical output at the ankle. This was confirmed by individual joint kinetic and biomechanical plasticity ratio differences between old and young adults. These findings confirm that age-associated biomechanical plasticity existed within our old cohort during all four walking conditions.

During level walking at self-selected speeds, statistically significant positive correlations were observed between physical capacity measures and biomechanical plasticity ratios. This indicates higher magnitudes of biomechanical plasticity in more capable old adults during level walking at self-selected speeds. During level walking at the controlled speed, only one of the eight correlation analyses (SF-36 PC scores with hip/ankle peak positive power ratio) between physical capacity measures and biomechanical plasticity ratios revealed a significant relationship.

During incline walking at self-selected speeds, significant positive relationships were observed for three of the eight correlation analyses between physical capacity measures and biomechanical plasticity ratios and two others bordered on significance ($r = 0.30$ and $r = 29$; critical value for these analyses being $r = 0.311$). This indicates higher magnitudes of biomechanical plasticity in more capable old adults during incline walking at self-selected speeds. During inclined walking at the controlled speed, no significant relationships were observed between physical capacity measures and biomechanical plasticity ratios.

Comparisons of level to incline walking conditions did not support our second hypothesis. Comparison of the correlation analyses conducted for level and incline walking at self-selected speeds weakly suggests that the relationship between physical capacity and biomechanical plasticity is more observable during level walking, however these data are not conclusive. The same is true for level compared to incline walking at the controlled speeds of 1.30 ms^{-1} and 1.20 ms^{-1}, respectively.

Chapter V: Discussion

The purpose of this thesis was to examine the relationships between physical capacity and biomechanical plasticity in old adults during level and incline walking. We hypothesized that, as physical capacity declines in old adults, age-associated biomechanical plasticity becomes more pronounced. We also hypothesized that the magnitude of biomechanical plasticity becomes more pronounced during the more challenging task of incline compared to level walking. In other words, the increase in magnitude of biomechanical plasticity per unit change of physical capacity will be higher during incline compared to level walking.

This chapter is divided into the following sections: development of the hypothesis, confirming age-associated biomechanical plasticity, examining the range of physical capacities in our old adult sample, relationships between physical capacity and biomechanical plasticity during level walking, relationships between physical capacity and biomechanical plasticity during incline walking, limitations, and a summary.

Development of the hypothesis

Compared to healthy young adults, healthy old adults exhibit increased hip and decreased ankle joint kinetics during the support phase of level walking [13,24,56,92,102,115]. This distal-to-proximal redistribution of joint torques and powers has been termed biomechanical plasticity and represents underlying neuromuscular adaptations made by old adults[24]. It is thought that changes to distal structures such as the triceps surae muscles and the Achilles tendon compromise the amount of torque and power generated at the ankle joints of old adults[35]. To counter reduced torque and power production at the ankle joint, healthy old adults appear to increase contributions from more proximal muscles – primarily the extensor muscles of the hip joint. That old adults are capable of increasing mechanical output from the hip joint muscles suggests the interesting possibility that these muscles do not experience the same age-related decline that the plantarflexors experience (i.e., asymmetric aging among different tissues). Age-associated biomechanical plasticity has also been observed in old adults during incline walking. However, whereas young adults increase both hip and ankle joint torques during incline walking, old adults increase hip but not ankle joint torques, suggesting a more pronounced biomechanical plasticity[37]. Need for larger propulsive GRFs[74] and the natural adaptation of increased output at the hip[70] during inclined walking may be two of the primary causes of increased biomechanical plasticity during incline compared to level walking.

Age-associated biomechanical plasticity is well-reported in *healthy* old adults. However, physical capacity (i.e., walking performance), which varies greatly in this population, might also have an impact on biomechanical plasticity. For example, Graf et al. (2005) reported that low-performance compared to healthy old adults exhibited increased hip extensor and decreased ankle plantarflexor peak positive powers[52]. Similarly, Hortobagyi et al. (2016) reported 17% larger relative hip and 17% lower relative ankle joint work in weak compared to strong old adults[56]. Finally, Buddhadev & Martin (2016) reported increased hip extensor and decreased ankle plantaflexor relative joint work in sedentary compared to active adults, however their results did not quite reach statistical significance (both $p < 0.10$)[13]. Combined, these comparison studies suggest that low compared to high capacity old adults exhibit larger magnitudes of biomechanical plasticity during locomotion; i.e., they have greater relative contribution from hip extensor muscles and a smaller relative contribution from ankle plantarflexor muscles. However, the precise relationship between physical capacity and biomechanical plasticity in old adults is unclear. That is, we do not currently understand how the magnitude of biomechanical plasticity varies across the physical capacity spectrum in old adults. Understanding the relationship between physical capacity and biomechanical plasticity in old adults may illuminate the underlying neuromuscular adaptations made with age and ultimately be used to develop interventions for increasing walking performance, reducing disability, and increasing overall quality of life. Therefore, the purpose of this study was to examine the relationships between physical capacity and age-associated biomechanical plasticity during level and incline walking. We hypothesized that as physical capacity declines, age-associated biomechanical plasticity would increase in magnitude. That is, as physical capacity declined, we expected to observe increased hip extensor relative to decreased ankle plantarflexor torques and powers. It was also hypothesized that the magnitude of biomechanical plasticity would become more pronounced during the more challenging task of incline compared to level walking. In other words, the increase in magnitude of biomechanical plasticity per unit change of physical capacity would be higher during incline compared to level walking.

Confirming age-associated biomechanical plasticity in old compared to young adults

Age-associated biomechanical plasticity, defined by increased hip extensor and decreased ankle plantarflexor contributions to support phase gait mechanics, represents an adaptation made by old adults during level and incline walking. Comparisons between our young and old participants confirmed that age-associated biomechanical plasticity existed during all four walking conditions tested: level and incline at both self-selected and controlled (level: 1.30 m/s; incline: 1.20 m/s) speeds.

During level walking at self-selected speeds, old adults exhibited larger relative hip and lower relative ankle positive joint work compared to young adults. Our results match closely with previously described samples of old adults. Specifically, our old adults contributed ~40%

and ~49% while the old adults described by DeVita & Hortobagyi (2000) contributed ~44% and ~51% to total positive joint work from the hip and ankle, respectively[24]. Compared to the young adult sample described by DeVita & Hortobagyi (2000), our young adults exhibited larger relative hip (~29% versus 16%) and lower relative ankle (57% versus 73%) joint contributions to total positive work, despite walking at similar speeds (1.49 m/s versus 1.48 m/s)[24]. These differences in our young compared to previously described young adults technically made it more difficult to observe biomechanical plasticity in our old adults. Despite this, we still managed to observe significantly larger relative hip and lower relative ankle in our old compared to young adults – that is, we still observed biomechanical plasticity.

To control for possible walking speed effects, we included a level walking condition at a controlled speed of 1.20 m/s. During the controlled speed condition, old adults exhibited larger relative hip and lower relative ankle positive joint work compared to young adults. Specifically, old adults contributed ~41% and ~49% while young adults contributed ~31% and ~57% to total positive joint work from the hip and ankle, respectively. The existence of biomechanical plasticity during the controlled speed condition confirms that this gait adaptation is not a consequence of decreased walking speed in old adults but rather, represents a fundamental neuromuscular adaptation that occurs with age[24,37,56]. Within both young and old adult samples, relative hip and ankle joint work were similar during self-selected and controlled speed conditions. This is consistent with at least one previous report of unchanging relative joint work values across slow, comfortable, and fast walking speeds[9]. Together, these results suggest that contributions from individual lower limb joints to total positive work do not change greatly across a variety of walking speeds. However, it should also be noted that others have reported changes in relative hip and knee, but not ankle joint contributions to total positive joint work while walking through a range (1.1 m/s to 1.7 m/s) of walking speeds[13]. It is possible that within-sample variability accounts for these differences.

Our comparisons of old versus young adults walking over level ground confirm the existence of age-associated biomechanical plasticity in our sample. These results are consistent with previous studies demonstrating biomechanical plasticity in old adults during level walking[13,24]. Disproportionate loss of muscle function in the plantarflexor muscles[15], decreased hip extension range of motion[64], changes to Achilles tendon dynamics[42], and balance impairments[63] in old adults are all factors that might contribute to biomechanical plasticity. Changes to medial gastrocnemius muscle architecture including decreased anatomic and physiological cross-sectional area and muscle volume may impair force production and result in decreased mechanical output at the ankle joint[80]. Indeed, decreased physiological cross-sectional area of the medial gastrocnemius appears to negatively influence torque and power production at the ankle joint[107]. Additionally, decreased mitochondrial function[57] and neuronal sprouting[88] in distal compared to proximal muscles might also contribute to decreased plantarflexor muscle function. Some have argued that decreased hip extension due to hip flexor "tightness" (hip contracture) in old adults causes a shorter stride length and in turn, reduced mechanical output from the ankle joint[64]. However, a 10-week hip flexor stretching protocol failed to have any effect on ankle joint mechanics during walking[111]. Age-related changes to the Achilles tendon including increased free tendon compliance[19,104] and higher degrees of sub-tendon coupling[42] may also negatively impact mechanical output from the ankle joint during locomotion. For example, increased tendon compliance likely reduces the amount of elastic energy stored by the Achilles tendon and reduces overall joint power production at the ankle. Because the Achilles tendon passively converts a large amount of stored elastic energy into mechanical work during locomotion[101], any factor that limits its ability to do so would have negative consequences for power generation at the ankle and likely increase total metabolic cost of walking. Indeed, limiting the ankle joint in healthy young adults using a brace caused a redistribution of mechanical work to the hip joint and increased metabolic power of walking by 7.4%[117]. Finally, it is possible that balance impairments in old adults may contribute to decreased ankle joint mechanical output. Kerrigan et al. reported 22% lower peak ankle power during push-off in old adults with a history of falls compared to old adults without a history of falls[63]. Additionally, old adults have been shown to rely more heavily on visual feedback during walking, suggesting the importance of sensory decline as well as muscular decline when considering biomechanical adaptations[36]. Although this is not a comprehensive list, the above-mentioned age-related changes are all possible contributing factors to lower torque and power production of the ankle plantarflexor muscles of old adults during locomotor tasks. Indeed, decreased ankle joint torque and power has been observed in level walking at self-selected[59,64,92,100], controlled[13,24,56], and fast[59] speeds as well as during running[22]. To compensate for decreased mechanical output at the ankle, old adults redistribute mechanical output to the more proximal hip joint. DeVita & Hortobagyi (2000) described this redistribution as a representation of the ability of the human body to undergo neuromuscular adaptations in the face of decreased function in a particular muscle group or joint[24]. Because old adults redistribute mechanical output to the hip during locomotion, it appears that the muscles and other tissues crossing this more proximal joint are more optimally suited for torque and power production during walking. If true, this suggests the interesting possibility that tissues in the human body do not all age at the same rate (i.e., tissues crossing the hip do not age at the same rate that tissues crossing the ankle age). Franz (2016) also makes the compelling case that sensorimotor decline and loss of sensory feedback from the ankle plantarflexors may cause old adults to rely more heavily on muscles crossing the hip – which are under feedforward control[35]. This, Franz (2016) suggests, may be a safer or more stable gait pattern for old adults[35].

During incline walking at self-selected speeds, old adults exhibited larger relative hip and lower relative ankle positive joint work compared to young adults. Specifically, our old adults contributed ~46% and ~34% while young adults contributed ~38% and ~41% of total positive joint work from the hip and ankle, respectively. During incline walking at the controlled speed, which was similar to the self-selected speed of old (self-selected: 1.23 m/s; controlled: 1.24 m/s) but not young (self-selected: 1.35 m/s; controlled: 1.20 m/s) adults, old adults contributed ~46% and ~34% while young adults contributed ~37% and ~42% of total positive joint work from the hip and ankle, respectively. These results confirm the existence of biomechanical plasticity in our sample of old adults during incline walking.

Incline walking requires that net-positive work be done to raise the body's center of mass with each step. To achieve this, healthy young adults perform positive work with both the trailing and leading limbs during double support while walking uphill[41]. This is different from level walking, where the lead leg performs negative work during double support. The underlying joint mechanics of incline gait reveal that healthy young adults rely primarily on increased hip extensor torque and positive power in early support and, to a lesser extent, increased ankle plantarflexor torque but not positive power in late support[37,70]. Compared to young adults, old adults exhibit 25% smaller ankle plantarflexor peak torques and 18% smaller ankle plantarflexor peak positive powers while walking at a 9° incline[37]. We report very similar findings while old adults walked at a 10° incline; our old adults exhibited ~25% and ~20% smaller ankle plantarflexor peak positive powers while walking at self-selected and controlled speeds, respectively, supporting the work of Franz & Kram (2014). We now expand the work of Franz & Kram (2014)[37] by reporting decreased relative ankle joint positive work in old compared to young adults during incline walking. This decreased mechanical output at the ankle joint is likely the cause of the decreased propulsive GRF and total trail leg positive work noted in old adults during incline walking[38]. Interestingly, rather than increasing hip extensor mechanical output to compensate for this ankle power deficit, Franz & Kram (2014) reported that old adults generated 119% larger peak hip flexor positive power immediately preceding toe-off[37], likely as a means of "pulling" the trailing leg into swing phase. Our data also show no differences in hip extensor peak positive powers between young and old adults during incline walking at either self-selected or controlled speeds. We do, however, report 7.4% and 9.5% larger relative hip extensor positive work in old compared to young adults during incline walking at self-selected and controlled speeds, respectively. It is possible that our old adults increased hip extensor mechanical output during early support to generate more positive power from the leading leg as a response to decreased mechanical output from the ankle joint of the contralateral (trailing) leg, however we did not collect bilateral data and cannot substantiate this claim. It is also possible that methodological differences account for these disparities in results. For example, Franz & Kram (2014) collected all gait data as participants walked on an instrumented treadmill while our gait data were collected while participants walked up a relatively short, custom built incline. Interestingly, although their joint-level mechanical data did not show increased hip extensor output, Franz & Kram (2013) have also supported the existence of age-associated biomechanical plasticity, as we've defined it, during incline walking with EMG data, reporting greater gluteus maximus and lower medial gastrocnemius activity at increasingly steep uphill grades[39]. Our results support their EMG data with joint-level mechanical data and confirm the existence of biomechanical plasticity in old adults during incline walking.

It seems likely that the age-related adaptations thought to contribute to biomechanical plasticity in old adults during level walking also cause plasticity during incline walking. However, because even healthy young adults rely primarily on increased mechanical output from the hip and not as much from the ankle joint during incline gait, it is difficult to determine how impairments at the ankle might impact incline walking performance. Because incline walking requires the generation of larger propulsive forces, and because the ankle joint plays a pivotal role in generating propulsive forces during locomotion, it follows that any impairment at the ankle might limit incline locomotion. In theory, any mode of locomotion that requires increased propulsive forces might exacerbate ankle impairments in old adults and cause increased magnitudes of biomechanical plasticity. This is supported by Franz & Kram (2014) who report that young adults increase both hip and ankle joint torques while old adults increase hip but not ankle joint torques during incline walking[37]. This idea partially guided our second hypothesis, however, the current analysis was strictly concerned with demonstrating the existence of biomechanical plasticity within our old adult sample and did not test magnitudes of biomechanical plasticity between level and incline gait.

Our comparisons of young and old adults confirmed the existence of biomechanical plasticity in our sample of old adults during all four walking conditions included in this study. Our level walking data matched well with previously described samples of young and old adults and confirmed that aging causes a distal-to-proximal redistribution of joint torques and powers. Our incline walking data support that the same age-associated biomechanical plasticity exhibited by old adults walking over level ground exists during incline walking.

Examining the range of physical capacities in our old adult sample

To properly test our first hypothesis, it was necessary to use a valid and reliable measure of physical capacity that could create measures or scores across a large range of values. We used Short-Form Health Survey Physical Component (SF-36 PC) scores and 20-meter level self-selected speeds.

The SF-36 is a self-report questionnaire meant to survey health based on eight broad categories, each with their own set of subcategories[110]. The SF-36 has been used to show physical capacity (quality of life) differences between highly active and inactive old adults[2]. Additionally, SF-36 PC scores have shown significant correlations with BMI, body strength (via arm curl test), walking endurance (via 6-minute walking test), and physical activity level[51]. Thus, the SF-36 PC score serves as a valid metric for physical capacity in old populations. Within our old population, the mean SF-36 PC score was 53.7 ± 6.2 with a range of 36.7 – 61.6. Wood et al. (2005) reported average SF-36 PC scores in high, moderate, and low functional fitness groups of old men and women. After averaging their men and women means, the following scores emerge: “low” = 40.7, “moderate” = 48.0, and “high” = 51.5[116]. Based on these averages, our group mean would be considered “high.” However, the average SF-36 PC score of our ten lowest scoring individuals was 43.0 ± 5.2 and the average score of our ten highest scoring individuals was 56.4 ± 1.3, indicating that we included individuals of both low and high physical capacities, based on SF-36 PC scores. Additionally, the averages reported by Wood et al. (2005) come from a study sample with a relatively high prevalence of cardiovascular disease, neurological disorders, orthopedic problems, and other diseases such as cancer[116]. The disease and disorder prevalence in their study population likely lowered their low, moderate, and high group averages for SF-36

PC scores. Because our study criteria excluded individuals with many of these diseases and disorders, it is not surprising that our study sample group mean might be considered “high.”

To include a more performance-based measure of physical capacity, we measured 20-meter level self-selected walking speeds of each participant. Self-selected walking speed in old adults has been associated with numerous adverse outcomes including falls, hospitalization, and even mortality[1,82,105]. Our study population had a group mean 20-meter self-selected speed of 1.26 ± 0.19 m/s and a range of 0.79 – 1.56 m/s. The average speed of our ten fastest participants was 1.46 ± 0.07 m/s and the average speed of our ten slowest participants was 1.05 ± 0. 12 m/s. These speeds are relatively consistent with previous studies reporting average gait speeds of 1.33 m/s in healthy[11] and 1.03 – 1.07 m/s[52,72] in low-capacity old adults. Although the mean speed of our ten lowest capacity (based on 20-meter self-selected speed) matches closely with these previous reports, the mean speed of our ten highest capacity old adults (based on 20-meter self-selected speed) is well above 1.33 m/s, indicating that we included some very high capacity old adults. Overall, based on 20-meter self-selected walking speeds, it appears that our old adult sample included individuals of low and very high physical capacities.

After comparing the physical capacity results of our sample to results of previous studies, we concluded that a sufficient capacity range of individuals existed in our sample and that the tools we used to measure physical capacity captured this. Having a sufficient range of physical capacity is important because these measures (SF-36 PC scores and 20-meter self-selected walking speeds) served as explanatory variables in correlation analyses.

Relationships between physical capacity and biomechanical plasticity during level walking

Previous comparisons of high versus low-capacity old adults suggested that low-capacity old adults exhibit larger magnitudes of biomechanical plasticity. Based on these comparison studies, we hypothesized that an inverse relationship would exist between physical capacity and biomechanical plasticity. Specifically, as physical capacity declined, we expected to observe increased hip extensor kinetics during early support relative to decreased ankle plantarflexor kinetics during late support. Our data did not support this hypothesis. In fact, our data suggested that a positive relationship exists between physical capacity and biomechanical plasticity during level walking.

During level walking at self-selected speeds, positive relationships were observed between SF-36 PC scores and all four biomechanical plasticity ratios. Additionally, a positive relationship was observed between SF-36 PC scores and relative hip extensor positive work and an inverse relationship was observed between SF-36 PC scores and relative ankle plantarflexor positive work. Using 20-meter self-selected walking speed as the measure of physical capacity yielded similar results. Specifically, positive relationships were observed between 20-meter self-selected walking speeds and three of the four biomechanical plasticity ratios. Additionally, an inverse relationship was observed between 20-meter self-selected walking speed and relative ankle plantarflexor positive work. These results indicate that physical capacity and biomechanical plasticity in old adults share a positive relationship during level walking at self-selected speeds. That is, old adults of higher physical capacities exhibited larger hip extensor and lower ankle plantarflexor mechanical output during this walking condition.

Similar relationships were observed during level walking at the controlled speed, however many of the correlation results did not reach statistical significance. For example, positive relationships existed between SF-36 PC scores and all four biomechanical plasticity ratios (r-values of 0.25 – 0.43), however only one of these relationships reached statistical significance. Using 20-meter self-selected walking speed as the measure of physical capacity yielded similar results – positive relationships between physical capacity and biomechanical plasticity ratios - none of the relationships reached statistical significance (r-values of 0.10 – 0.27). These results might weakly suggest that physical capacity and biomechanical plasticity also share a positive relationship when walking speed is controlled, however the lack of statistically significant correlations does not provide strong support for this conclusion.

Although our results do not support our hypothesis and contradict the studies upon which we built our hypothesis, they are consistent with a few previous studies. For example, our results are consistent with a previous comparison of active and inactive old adults. Savelberg et al. (2007) reported increased hip extensor and decreased ankle plantarflexor torques in old adult runners compared to inactive old adults, however their results did not reach statistical significance[100]. It is possible that larger sample sizes would have pushed these differences to statistical significance. More recently, Beijersbergen et al. (2016) reported increased hip extensor and decreased ankle plantaflexor mechanical output in old adults following a 10-week power training protocol[9]. Their results suggest that increasing capacity in old adults results in increased magnitudes of biomechanical plasticity. McGibbon & Krebs (2004) attempted to decouple aging effects from impairment effects to gait in old adults. In doing so, these authors reported increased hip extensor positive power as a discriminatory variable between healthy old and young and increased hip flexor negative power as a discriminatory variable between healthy old and impaired old[71]. These results suggest that, although *all* old adults exhibit a general shift towards more proximal muscles, *healthy* old appear to increase active contributions from the hip extensors while *impaired* old appear to increase passive action of the hip flexors. Our data are similar in that hip extensor kinetics shared a positive relationship with physical capacity (i.e., increased hip extensor activity in healthier old adults). To be completely consistent with McGibbon & Krebs (2004), we would expect to see an inverse relationship between physical capacity and hip flexor kinetics during mid to late support. Because our purpose was to explore biomechanical plasticity - defined here as a redistribution to hip extensor active contributions in early support – we did not conduct analyses on hip flexor kinetics in mid or late support. McGibbon & Krebs (2004) suggested that passive action of the hip flexors in mid-support and into late-support might act to propel the trunk and pelvis forward in a manner that increases whole body momentum[71]. Passive/eccentric compared to active/concentric muscle action is also more metabolically and energetically efficient[65,99]. It is possible that impaired or low-capacity old adults rely more heavily on passive muscular contributions during walking as a means of decreasing metabolic cost while healthy or high capacity old adults rely more heavily on active muscular action of the hip extensors as a means of increasing walking speed. It seems more likely, however, that low-capacity old adults have simply lost a greater amount of their ability to produce muscular force via concentric contraction, not only with the plantarflexors, but also with the hip extensors. Loss of function or force-producing capacity of the hip extensors would force low capacity old adults to either slow down or adapt a new gait pattern – such as increasing passive contributions from the hip flexor muscles in mid and late support.

It is possible that natural, healthy aging alters different components of the human body such as the material and structural properties of soft tissues and sensory/sensorsimotor function in old adults. In the face of these detrimental alterations, it appears that higher capacity old adults maintain an ability to undergo biomechanical adaptations. This is likely what age-associated biomechanical plasticity represents – the ability of healthy old adults to neuromechanically redistribute output from the ankle joint to the hip joint. This redistribution most likely affords some functional benefit(s). It is possible that age-associated biomechanical plasticity is a mechanism for increasing walking performance in old adults. For example, two of the four biomechanical plasticity ratios were positively related to in-trial self-selected level walking speed and three ratios were positively related to stride frequency while no relationships existed between plasticity and stride length, in our old adults (**Table 11**). This suggests that increased magnitudes of plasticity allow old adults of higher physical capacities to walk at faster comfortable speeds. Further, the positive relationship between plasticity and stride frequency, but not stride length, at self-selected speeds may suggest that biomechanical plasticity serves to increase gait in a safer manner in higher capacity old adults. Here, I assume that increased stride frequency results in a "safer" gait because it likely leads to increased double support time, and having both feet on the ground for a longer period of time increases the base of support for the body's center of mass for a longer period of time. Additionally, as discussed earlier, it is possible that age-related changes to sensorimotor function causes old adults to shift reliance to more proximal muscles in order to ensure stability during locomotion[35].

Table 11: Pearson product-moment correlation coefficients (r-value) between biomechanical plasticity ratios and walking speeds, stride lengths, and stride frequencies during the level walking at self-selected speed trials. Significant r-values are reported in **bold** print.

Biomechanical Plasticity and Self-Selected Spatiotemporal Variables During Level Walking			
Hip-to-Ankle Ratios	r-values		
	Walking Speed	Stride Length	Stride Frequency
Peak Torque	**0.518**	0.152	**0.614**
Angular Impulse	**0.353**	0.020	**0.514**
Peak Positive Power	0.252	-0.017	**0.412**
Positive Work	0.087	-0.124	0.290

It must be noted that our results are not analogous with the studies upon which we built our first hypothesis. Graf et al. (2005) reported increased hip extensor and decreased ankle plantarflexor peak positive powers in low-performance compared to healthy old adults walking at comfortable speeds[52]. Many of the low-performance participants included by Graf et al. (2005) exhibited health conditions that may have themselves altered gait mechanics independent of age-related changes. Specifically, 44% of their low-performance sample reported one or more falls in the year prior to testing whereas present participants did not have falls in this time period. It is possible that individuals who have a history of falls, but are otherwise relatively healthy, make gait adaptations to increase stability and reduce the likelihood of future falls. Kerrigan et al. (2000) reported that otherwise healthy old adults with a history of falls exhibited increased hip and decreased ankle mechanical output, suggesting that old fallers compared to non-fallers exhibited larger magnitudes of biomechanical plasticity[63]. Of the 12 fallers included by Graf et al. (2005), 9 reported “good” or “excellent” self-health[52]. Fallers included by both Graf et al. (2005) and Kerrigan et al. (2000) may have been healthy enough to incorporate biomechanical plasticity in response to a previous fall. If this were the case, their results would partially support our conclusion – that biomechanical plasticity represents a gait adaptation made by relatively high capacity old adults in order to increase walking performance. Additionally, some of the low-performance individuals included by Graf et al. (2005) had one or more joint replacements[52] whereas none of our participants had any joint replacement. It is possible that gait adaptations in the low-performance group reported by Graf et al. (2005) were partially prosthetic-driven rather than age-driven. More recently, Hortobagyi et al. (2016) reported that weak compared to strong old adults contributed 17% more work from the hip and 17% less work from the ankle plantarflexors during level walking[56]. Although leg strength has been previously associated with mobility decline[12], others have shown that muscular power is a more valid predictor of mobility in old adults[6,76]. This may partially explain the increase in biomechanical plasticity in old adults following a 10-week power training protocol[9]. It is also possible that factors other than muscular strength and power, such as sensorimotor decline and balance impairment, are larger contributors to gait adaptations in old adults. Finally, Buddhadev & Martin (2016) reported larger relative hip extensor and lower relative ankle plantarflexor joint work in sedentary compared to active adults, however these results did not reach statistical significance ($p < 0.10$)[13]. It is important to note that their comparison of active and sedentary individuals pooled both young and old adult samples together. Additionally, although no longitudinal study has tested how biomechanical plasticity progresses over time, gradual changes in underlying biological functions suggest that mechanical changes also happen gradually over time. Therefore, an individual's current physical activity level might not accurately reflect or predict the magnitude of his/her biomechanical adaptation.

Relationships between physical capacity and biomechanical plasticity during incline walking

Although, to my knowledge, no comparison of high and low capacity old adults during incline walking exists in the literature, differences between healthy young and old adults suggested increased magnitudes of biomechanical plasticity during the more difficult task of incline compared to level walking[37]. Based on previous comparisons of healthy young and old adults, we hypothesized that the increase in magnitude of biomechanical plasticity per unit change of physical capacity will be higher during incline compared to level walking. Based on our mixed correlation results during incline walking conditions, and significant overlap between 95% C.I.s for the beta weights of the few corresponding significant relationships that existed between level and incline walking conditions, we must reject this hypothesis.

During incline walking at self-selected speeds, we observed significant positive relationships between SF-36 PC scores and three of the four biomechanical plasticity ratios. We also observed positive relationships between SF-36 PC scores and many of hip extensor kinetic variables of interest, however relationships with ankle plantarflexor kinetics varied (some positive and others negative, but many non-significant). While using 20-meter level self-selected speeds as our measure of physical capacity in this condition's analyses, very few relationships reached statistical significance. During incline walking at the controlled speed, very few relationships reached statistical significance. This was the case while using both SF-36 PC scores and 20-meter self-selected speeds as predictor variables (physical capacity measures) in correlation analyses. Comparisons of the 95% C.I.s of the beta weights of corresponding significant relationships during level and incline walking at self-selected and controlled speeds revealed large overlaps. These comparisons suggest that the magnitude of biomechanical plasticity per unit change of physical capacity is not larger during incline compared to level walking.

Compared to level walking, incline walking requires the generation of larger propulsive GRFs[74]. Compared to young adults, old adults generate smaller propulsive GRFs during locomotion[38,43]. Franz (2016) provided an in-depth review of possible causes for this limitation in old adults[35], some of which were described in the discussion of our young versus old comparisons. Because mechanical output from the ankle joint is thought to contribute largely to generating propulsive GRFs, it follows that any ankle impairment might limit one's ability to walk over inclined surfaces. Our young versus old adult comparisons confirmed that old adults exhibited smaller relative ankle joint positive work. However we did not observe significant relationships between either physical capacity measure and relative ankle joint work. It is possible that our measures of physical capacity were simply not strong predictors of ankle joint mechanics in old adults during incline walking. It is also possible that some of the old adults in our sample relied more heavily on increased hip flexor rather than extensor muscles during incline walking, similar the old adult sample described by Franz & Kram (2014)[37]. Interestingly, many significant positive relationships were observed between SF-36 PC scores and hip extensor kinetics, particularly during incline walking at self-selected speeds. This may indicate that old adults of higher capacities maintain an ability to increase hip extensor mechanical output in order to walk over inclined surfaces. Perhaps high capacity old adults have maintained more concentric strength of the hip extensors compared to the low capacity old adults, however we did not include any measurement of strength in our protocol.

During incline walking, we observed significant relationships between physical capacity and biomechanical plasticity while using SF-36 PC scores, but not 20-meter self-selected walking speeds, as the explanatory variables in correlation analyses. It is possible that 20-meter self-selected walking speed does not accurately predict biomechanical plasticity during incline walking because incline walking requires a unique gait strategy. Thus, increased performance in one gait may not directly transfer to increased performance in the other. It is also possible that our self-reported measure of physical capacity (SF-36 PC score) was more indicative of each individual's self-confidence, which may have served as a more robust predictor of biomechanical plasticity during incline walking. For example, the ramp used for incline gait analyses in our lab is short in length, has a large incline angle, and does not have handrails or a harness in place to reduce fall risk. These factors may have induced a fear of falling and subsequent gait changes in our old adults. The SF-36 PC score may have more accurately captured the possibility or magnitude of this fear and thus, served as a stronger predictor of biomechanical plasticity. We did, however, seek to eliminate fear of falling during the incline condition by allowing all individuals practice trials on the first and second days of testing prior to collecting any gait data.

Limitations

The current protocol has several limitations in design that reduce the generalizability of the work. Biomechanical data from our sample was collected on only the pelvis and right leg of each participant and we assumed bilateral symmetry when discussing our results. The incline condition included in this study was conducted over a relatively short ramp. Although the ramp was long enough to allow at least one footfall of the right and left foot prior to contact with the force platform, it is possible that this was not enough time to "adapt" an uphill walking pattern. This may explain, at least partially, differences between our data and previous reports of incline walking in old adults – particularly those of Franz and colleagues, who conducted incline gait analyses using an instrumented treadmill that allowed for longer continuous walking trials. However, kinematic adaptations following the first step from a level to an incline surface suggests that individuals adapt biomechanically within this first step[89]. Finally, it is possible that age has a more profound effect on level gait mechanics than physical capacity. Buddhadev & Martin (2016) recently reported that age, but not physical activity status, was the primary cause of proximal work redistribution, however they pooled both young and old adults for this analysis[13]. Although many of our observed correlations between biomechanical plasticity and physical capacity were statistically significant, the coefficients of determination (r^2) were relatively weak. This suggests physical capacity, as we defined it, might be a relatively weak predictor of biomechanical plasticity in old adults.

Summary

Our comparisons of old and young adults confirmed the existence of biomechanical plasticity in our old adult sample during both level and incline walking. Compared to our young, our old adults exhibited larger relative hip extensor and smaller relative ankle plantarflexor work while walking over level and incline surfaces at both self-selected and controlled (level: 1.30 m/s; incline: 1.20 m/s) speeds. Our results are consistent with previous comparisons of young and old adults and confirm that biomechanical plasticity represents a fundamental biomechanical adaptation with age.

Based on the results from our correlation analyses, we reject our first hypothesis that physical capacity and biomechanical plasticity would share an inverse relationship. In fact, during level walking at self-selected and controlled speeds, physical capacity and biomechanical plasticity in old adults shared a positive relationship. However, during level walking at the controlled speed, many of these relationships did not reach statistical significance. The positive relationships between biomechanical plasticity ratios and in-trial self-selected walking speeds might suggest that increased magnitudes of plasticity helped higher capacity old adults walk at faster speeds. Incline walking at self-selected and controlled speeds yielded mixed results.

Our data indicate that biomechanical plasticity represents a level walking gait adaptation made by higher capacity old adults that might afford functional benefits – specifically, it allows individuals to walk faster. Further, an inability to incorporate biomechanical plasticity might impair walking performance and contribute to declining capacity old adults. Based on these findings, we now propose that age-associated biomechanical plasticity is representative of a robustness to adaptation in higher capacity old adults. If this is true, we should not seek to reduce the magnitude of biomechanical plasticity (i.e., make joint-level mechanics in old adults more similar to young adults) in old adults by implementing training interventions targeting the ankle plantarflexors. Rather, we should attempt to increase the recruitment and quality of hip joint musculature in old adults. Such interventions might result in higher magnitudes of biomechanical plasticity and increased walking performance in old adult populations. Results from our cross-sectional design may provide the framework for a longitudinal intervention study aimed at increasing biomechanical plasticity and thereby walking performance in old adults. Increased walking performance in this population has the potential to decrease adverse outcomes such as falls, hospitalizations, and even mortality, leading to an overall increased quality of life.

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Appendix A: Institutional Review Board Approval

EAST CAROLINA UNIVERSITY
University & Medical Center Institutional Review Board Office
4N-70 Brody Medical Sciences Building· Mail Stop 682
600 Moye Boulevard · Greenville, NC 27834
Office 252-744-2914 · Fax 252-744-2284 · www.ecu.edu/irb

Notification of Initial Approval: Expedited

From: Biomedical IRB
To: Paul DeVita
CC:
Date: 3/10/2016
Re: UMCIRB 16-000055
Biomechanical plasticity and physical capacity

I am pleased to inform you that your Expedited Application was approved. Approval of the study and any consent form(s) is for the period of 3/8/2016 to 3/7/2017. The research study is eligible for review under expedited category #4,7. The Chairperson (or designee) deemed this study no more than minimal risk.

Changes to this approved research may not be initiated without UMCIRB review except when necessary to eliminate an apparent immediate hazard to the participant. All unanticipated problems involving risks to participants and others must be promptly reported to the UMCIRB. The investigator must submit a continuing review/closure application to the UMCIRB prior to the date of study expiration. The Investigator must adhere to all reporting requirements for this study.

Approved consent documents with the IRB approval date stamped on the document should be used to consent participants (consent documents with the IRB approval date stamp are found under the Documents tab in the study workspace).

The approval includes the following items:

Name	Description
Physical capacity & plasticity - informed consent 2016.docx	Consent Forms
Physical capacity & plasticity - online announcement for recruitment 2016.pptx	Recruitment Documents/Scripts
Physical capacity & plasticity - print announcement for recruitment 2016.pdf	Recruitment Documents/Scripts
Physical capacity & plasticity - screener.docx	Surveys and Questionnaires
Plasticity & Physical Capacity - Protocol.docx	Study Protocol or Grant Application
SF-36.pdf	Surveys and Questionnaires

The Chairperson (or designee) does not have a potential for conflict of interest on this study.

Appendix B: Informed Consent Form

East Carolina University

Informed Consent to Participate in Research

Information to consider before taking part in research that has no more than minimal risk.

Title of Research Study: The relationships between physical capacity and biomechanical plasticity with age during level and incline walking
Principal Investigator: Paul DeVita
Institution, Department or Division: Kinesiology
Address: 332 Ward Sports Medicine Building, East Carolina University
Telephone #: 252-737-4616

Researchers at East Carolina University (ECU) study issues related to society, health problems, environmental problems, behavior problems and the human condition. To do this, we need the help of volunteers who are willing to take part in research.

Why am I being invited to take part in this research?

The purpose of this research is to examine walking gait adaptations that occur in elderly adults of differing physical capacities. You are being invited to take part in this research because you meet the inclusion criteria and appear to be free of contraindications to participating in this study. The inclusion criteria for this study are: 18-25 years old or 70-85 years old, non-smoker, and able to perform regular daily activities such as walking, climbing stairs and inclines, and rising from a chair without assistance. 18-25 year old participants should also engage in regular physical activity (minimum of 3 times per week). By doing this research, we hope to learn more about walking gait adaptations that occur in elderly adults across a range of physical capacities.

If you volunteer to take part in this research, you will be one of about 40 people to do so.

Are there reasons I should not take part in this research?

I understand that I should not partake in this research if I do not meet the inclusion criteria, have had a musculoskeletal injury in the past 6 months, history of lower limb, back, or joint replacement surgery, neurological or neuromuscular disorder such as Parkinson's disease or stroke, cardiac disease, or any terminal illness, or use any tobacco products.

What other choices do I have if I do not take part in this research?

You can choose not to participate.

Where is the research going to take place and how long will it last?

The research will be conducted in the Biomechanics Laboratory, room 332 Ward Sports Medicine Building at East Carolina University. You will need to come to the Biomechanics Laboratory two separate times during the study. The total amount of time you will be asked to volunteer for this study is approximately 3 hours over these two visits.

What will I be asked to do?

You will be asked to do the following:

- Complete a short questionnaire that includes relevant demographic information as well as a short health history. This questionnaire is used to ensure participation eligibility.
- Complete the 36-Item-Short-Form Health Survey (SF-36), complete two 20-meter walks at a self-selected speed as well as a safe-maximal speed, and complete two walking trials on our incline ramp at a self-selected

speed. Complete a 6-minute walk test. The 6-minute walk test includes 6 minutes of continuous walking at a self-selected speed over a level, obstacle free surface. These tests will be used to help determine physical capacity. You will be asked to complete the 10-Item State Trait Anxiety Index following the two incline ramp walking trials.

- Undergo biomechanical gait analysis. This testing method will include walking over a level walkway and up an incline ramp (3.2-meters long; 10% incline) in the Lab. During the testing session, small spherical reflective markers will be placed on your pelvis and right leg.
- The total walking time for both Lab visits is estimated to be 1 hour and 40 minutes. You will be asked to walk approximately 25 minutes during the initial visit and 75 minutes on the final visit.

What might I experience if I take part in the research?
We don't know of any risks (the chance of harm) associated with this research. Any risks that may occur with this research are no more than what you would experience in everyday life. We don't know if you will benefit from taking part in this study. There may not be any personal benefit to you but the information gained by doing this research may help others in the future.

Will I be paid for taking part in this research?
We will be able to pay you $30 for the time you volunteer while being in this study. The $30 payment will be in the form of a gift card.

Will it cost me to take part in this research?
It will not cost you any money to be part of the research.

Who will know that I took part in this research and learn personal information about me?
ECU and the people and organizations listed below may know that you took part in this research and may see information about you that is normally kept private. With your permission, these people may use your private information to do this research:

- Any agency of the federal, state, or local government that regulates human research. This includes the Department of Health and Human Services (DHHS), the North Carolina Department of Health, and the Office for Human Research Protections.
- The University & Medical Center Institutional Review Board (UMCIRB) and its staff have responsibility for overseeing your welfare during this research and may need to see research records that identify you.
- Paul DeVita, the primary investigator and faculty supervisor, and Daniel Kuhman, the sub-investigator.

How will you keep the information you collect about me secure? How long will you keep it?
All data files will be kept for 5 years after the study is completed. The investigators will keep your personal data in strict confidence by having your data coded. Instead of your name, you will be identified in the data records with an alphanumeric identity number. Your name and identity number will not be identified in any subsequent report or publication. The members of our research team will be the only people who know the identity number associated with your name. Any files that associate your name with your identity number will be encrypted and only members of our research team will know the password to these files. The data collected during this study will be used for research purposes.

What if I decide I don't want to continue in this research?
You can stop at any time after it has already started. There will be no consequences if you stop and you will not be criticized. You will not lose any benefits that you normally receive.

Who should I contact if I have questions?

The people conducting this study will be able to answer any questions concerning this research, now or in the future. You may contact the Principal Investigator, Paul DeVita, at 252-737-4563 (work days, between 9am and 5pm) or the student investigator, Daniel Kuhman, at 252-737-4616 (work days, between 9am and 5pm).

If you have questions about your rights as someone taking part in research, you may call the Office of Research Integrity & Compliance (ORIC) at phone number 252-744-2914 (days, 8:00 am-5:00 pm). If you would like to report a complaint or concern about this research study, you may call the Director of the ORIC, at 252-744-1971.

I have decided I want to take part in this research. What should I do now?

The person obtaining informed consent will ask you to read the following and if you agree, you should sign this form:

- I have read (or had read to me) all of the above information.
- I have had an opportunity to ask questions about things in this research I did not understand and have received satisfactory answers.
- I know that I can stop taking part in this study at any time.
- By signing this informed consent form, I am not giving up any of my rights.
- I have been given a copy of this consent document, and it is mine to keep.

____

Participant's Name (PRINT) Signature Date

Person Obtaining Informed Consent: I have conducted the initial informed consent process. I have orally reviewed the contents of the consent document with the person who has signed above, and answered all of the person's questions about the research.

____

Person Obtaining Consent (PRINT) Signature Date

Appendix C: The Short-Form Health Survey (SF-36)

Short Form Health History Form

1) In general, would you say your health is (circle one):

Excellent Very Good Good Fair Poor

2) *Compared to one year ago*, how would you rate your health in general *now* (circle one)?

Much better now than one year ago

Somewhat better now than one year ago

About the same as one year ago

Somewhat worse now than one year ago

Much worse than one year ago

3) The following items are about activities you might do during a typical day. Does *your health now limit you* in these activities? If so, how much (circle one)?

a) *Vigorous activities*, such as running, lifting heavy objects, participating in strenuous sports.

Yes, Limited a lot Yes, Limited a little No, Not limited at all

b) *Moderate activities*, such as moving a table, pushing a vacuum cleaner, bowling, or playing golf.

Yes, Limited a lot Yes, Limited a little No, Not limited at all

c) Lifting or carrying groceries

Yes, Limited a lot Yes, Limited a little No, Not limited at all

d) Climbing *several* flights of stairs

Yes, Limited a lot Yes, Limited a little No, Not limited at all

e) Climbing *one* flight of stairs

Yes, Limited a lot Yes, Limited a little No, Not limited at all

f) Bending, kneeling, or stooping

Yes, Limited a lot Yes, Limited a little No, Not limited at all

g) Walking *more than a mile*

Yes, Limited a lot Yes, Limited a little No, Not limited at all

h) Walking *several blocks*

Yes, Limited a lot Yes, Limited a little No, Not limited at all

i) Walking *one block*

Yes, Limited a lot Yes, Limited a little No, Not limited at all

j) Bathing or dressing yourself

Yes, Limited a lot Yes, Limited a little No, Not limited at all

4) During the *past 4 weeks*, have you had any of the following problems with your work or other regular daily activities *as a result of your physical health*?

a) Cut down the *amount of time* you spent on work or other activities

Yes No

b) *Accomplished less* than you would like

Yes No

c) Were limited in the *kind* of work or other activities

Yes No

d) Had *difficulty* performing the work or other activities (for example, it took extra effort)

Yes No

5) During the *past 4 weeks*, have you had any of the following problems with your work or other regular daily activities *as a result of any emotional problems* (such as feeling depressed or anxious)?

a) Cut down the *amount of time* you spent on work or other activities

Yes No

b) *Accomplished less* than you would like

Yes No

c) Didn't do work or other activities *as carefully* as usual

Yes No

6) During the *past 4 weeks*, to what extent has your physical health or emotional problems interfered with your normal social activities with family, friends, neighbors, or groups (circle one)?

Not at all Slightly Moderately Quite a bit Extremely

7) How much *bodily* pain have you had during the *past 4 weeks* (circle one)?

None Very mild Mild Moderate Severe Very Severe

8) During the *past 4 weeks*, how much did *pain* interfere with your normal work (including both work outside the home and housework) (circle one)?

Not at all A little bit Moderately Quite a bit Extremely

9) These questions are about how you feel and how things have been with you *during the past 4 weeks*. For each question, please give the one answer that comes closest to the way you have been feeling. How much of the time during the *past 4 weeks*…

a) Did you feel full of pep?

All of the time	Most of the time
A good bit of the time	Some of the time
A little of the time	None of the time

b) Have you been a very nervous person?

All of the time	Most of the time
A good bit of the time	Some of the time
A little of the time	None of the time

c) Have you felt so down in the dumps that nothing could cheer you up?

All of the time	Most of the time
A good bit of the time	Some of the time
A little of the time	None of the time

d) Have you felt calm and peaceful?

All of the time	Most of the time
A good bit of the time	Some of the time
A little of the time	None of the time

e) Did you have a lot of energy?

All of the time	Most of the time
A good bit of the time	Some of the time
A little of the time	None of the time

f) Have you felt downhearted and blue?

All of the time	Most of the time
A good bit of the time	Some of the time
A little of the time	None of the time

g) Did you feel worn out?

All of the time	Most of the time
A good bit of the time	Some of the time
A little of the time	None of the time

h) Have you been a happy person?

All of the time	Most of the time
A good bit of the time	Some of the time
A little of the time	None of the time

i) Did you feel tired?

All of the time	Most of the time
A good bit of the time	Some of the time
A little of the time	None of the time

10) During the *past 4 weeks*, how much of the time has your *physical health or emotional problems* interfered with your social activities (like visiting with friends, relatives, etc.) (circle one)?

All of the time

Most of the time

Some of the time

A little of the time

None of the time

11) How TRUE or FALSE is *each* of the following statements for you (circle one)?

a) I seem to get sick a little easier than other people

Definitely true Mostly true Don't know Mostly false Definitely false

b) I am as healthy as anybody I know

Definitely true Mostly true Don't know Mostly false Definitely false

c) I expect my health to get worse

Definitely true Mostly true Don't know Mostly false Definitely false

d) My health is excellent

Definitely true Mostly true Don't know Mostly false Definitely false

Appendix D: Health Questionnaire

<u>**Health Questionnaire to Determine Eligibility for Research Participants**</u>

Demographic data: Date________________________

Name ________________________ Phone number________________________

Address____

Birth date ________________________ Age _________ Gender M F

Height (ft/in) ________________ Height (m)________________

Weight (lbs) ________________ Mass (kg)_________________

BMI (kg/m^2) ________________

Do you smoke? Yes____ No ____

Have you smoked in the past? Yes____ No ____

If yes, when did you stop smoking________________________

Functional ability in daily activities:

Are you able to leave your house on a daily basis without aid? Yes _____ No _____

Can you do the following activities independently?

Dress Yes____ No ____

Walk Yes____ No ____

Climb stairs Yes____ No ____

Rise from a chair Yes____ No ____

Do you use a walker or cane when walking? Yes____ No ____

During the past year, did you fall down more than once while walking or climbing stairs?
Yes____ No ___

What physical activities do you regularly perform (e.g. run, tennis, basketball)?

____

____

How often do you do these activities (3 days/week is minimum)?

____

____

Medical:

In the past 6 months, have you suffered any musculoskeletal injuries? Yes____ No ____

Do you have a history of joint replacement surgery in the lower limb? Yes____ No ____

Do you have osteoarthritis in any of the joints in your lower-limb? Yes ____ No ____

Do you have any neurological problems such as stroke or Parkinson's disease? Yes___ No___

Do you have any problems with your heart such as atrial fibrillation, pace maker, coronary artery disease, or congestive heart failure? Yes____ No ____

Do you have any pulmonary diseases such as difficulty in breathing or emphysema?
Yes____ No ____

Do you have any peripheral artery disease? Yes____ No ____

Do you have high blood pressure (>160/90 mm Hg)? Yes____ No ____

Do you take medication to control your blood pressure? Yes____ No_____

Have you ever been diagnosed with cancer? Yes _____ No _____

Do you have any loss of vision? Yes____ No ____

If yes, do you have eye glasses or contact lenses that correct your vision? Yes___ No___

Do you have any other medical problems we did not talk about? Yes____ No_____

If, "Yes," what is or are the conditions?

____

____

Please list any surgeries you have had.

____

____

Please tell us any other health illnesses you have had or currently have.

____

____