ABSTRACT

Title of Dissertation: INVESTIGATING SOURCES OF AGE-RELATED DIFFERENCES IN WALKING MECHANICS

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Walking is one of the most common activities of daily living and represents independence and improved quality of life, particularly among older adults. However, many older adults report substantial mobility challenges, which may be associated with age-related differences in lower-extremity gait kinetics. These differences are summarily referred to as a 'distal to proximal shift' of joint moments and powers, and are characterized by smaller ankle kinetics and larger hip kinetics in older vs. young adults. Although age-related differences in walking mechanics are well-documented, there is little consensus about which biomechanical factors contribute to these differences. Addressing this gap in knowledge is an important step in determining if this shift is preventable, or rather, an unavoidable part of healthy aging. Therefore, the overarching goal of this dissertation was to investigate sources of the age-related distal to proximal shift in gait kinetics. Specifically, this dissertation determined the extent to which the shift in kinetics is explained by age-related differences in (i) step length and trunk kinematics, (ii) years of endurance running (i.e., habitual physical activity), and (iii) gastrocnemius muscle architecture and individual lower-extremity muscle forces.

In study 1, step length and trunk position did not reverse or reduce the age-related distal to proximal shift. Similarly, in study 2, a history of habitual endurance running did not reduce or reverse the shift. The third study confirmed the distal to proximal shift at the muscle level, suggesting that gastrocnemius may be a primary site of age-related differences in plantarflexor force, due to the shorter gastrocnemius muscle fascicles and smaller gastrocnemius force production in older adults vs. young adults. The present findings support the notion that the age-related distal to proximal shift of kinetics in active older adults is due primarily to differences at the muscle level and do not support previous speculations that this shift is due to spatiotemporal factors such as step length, joint kinematics, or physical activity. Further, these results suggest that age-related differences in lower-extremity joint kinetics are an unavoidable part of natural aging even in the absence of mobility limitations and the presence of a lifelong history of endurance running.

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MECHANICS

by

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CHAPTER 1

INTRODUCTION

1.1 Background

Walking is one of the most common activities of daily living and the primary means of locomotion for many older adults. As such, maintaining the ability to walk is important for preserving independence and mobility in old age, yet one of every four adults over age 55 report difficulty walking one quarter of a mile (Schoenborn & Heyman, 2009). The mobility challenges faced by older adults may be related to or reflected by changes in walking mechanics that occur with aging. The overall goal of the present research is to better understand why older adults walk differently from young adults. This work is important because a better understanding of the mechanisms behind age-related differences in gait characteristics will serve as a starting point for future research seeking to improve mobility in older adults.

During walking, older adults display differences in lower limb joint mechanics compared to young adults. These changes have been referred to as a "distal to proximal shift" of lower-extremity joint moments and powers (DeVita & Hortobagyi, 2000b). Specifically, older adults exhibit smaller overall ankle joint moments and powers and greater hip joint moments and powers compared to young adults. Here "older adults" are typically defined as individuals ~60-65 years of age and older, with "young adults" defined as individuals ~18-35 years of age. This age-related distal to proximal shift has been reported in multiple studies (Cofré et al., 2011; DeVita & Hortobagyi, 2000) but its mechanisms are, to date, unknown. Potential explanations suggested by previous

observational studies on older adults include step length, trunk lean, and physical activity/fitness (Boyer et al., 2012) but none of these potential effects have been extensively examined, experimentally.

Age-related differences in step length are well-documented and posited as a potential mechanism for the distal to proximal shift. Older adults often take shorter steps during walking, even when walking at the same speed as young adults (DeVita & Hortobagyi, 2000b; Ko et al., 2009a; Winter et al., 1990). Walking with shorter step lengths is associated with a smaller ankle range of motion during stance and a smaller plantarflexion angle at pushoff (Judge et al., 1996a; Murray et al., 1969). Since joint moments are a product of inertia and angular acceleration (the second derivative of angular position), a smaller angular displacement would likely contribute to smaller ankle kinetics. Indeed, step length is a strong predictor of ankle joint power in older adults (Judge et al., 1996) which suggests the age-related distal to proximal shift may be due to characteristically shorter step lengths in older adults. Modifying step length in young adults resulted in changes to ankle mechanics such that shorter steps decreased ankle power, moment, and impulse (Umberger & Martin, 2007). Therefore, increasing step length in older adults may result in increased ankle kinetics, but this effect is unknown because there have been no systematic investigations of step length on gait kinetics in older adults.

Joint and segment kinematics may also contribute to age-related differences in joint kinetics. Older adults tend to walk with more hip flexion (Anderson & Madigan, 2014a; Anderson & Madigan, 2014b; Cofré et al., 2011) and less hip extension (i.e. older adults walk with a more "bent over" posture) (Anderson & Madigan, 2014a; Anderson &

Madigan, 2014b), compared to young adults. Since hip extensor muscles can produce more torque at larger flexion angles (Anderson et al., 2007), larger hip flexion due to forward trunk lean may result in increased hip extensor moments and powers during gait in older adults. There have been no systematic investigation of step length or trunk kinematics. Addressing these gaps in the literature will help elucidate mechanisms responsible for the age-related differences in joint kinetics. Further, understanding the influence of trunk position on hip kinetics in older adults will help characterize the role of the hip joint in the distal to proximal shift of kinetics.

A point of contention in the literature is whether these age-related differences in kinetics are at least partially due to age-related declines in physical activity. Regular physical activity helps to maintain muscle strength and mobility in older adults (Sandler et al., 1991; Visser et al., 2002), but physical activity levels often decline with age (Hallal et al., 2012; Milanović et al., 2013). Considering that leg strength partially mediates age-related differences in gait kinetics (Hortobágyi et al., 2016), and more specifically, that plantarflexor strength correlates with peak plantarflexor power during gait (Silder et al., 2008), it is possible that the age-related decrease in physical activity level is at least partially contributing to the age-related distal to proximal shift in gait kinetics through a combination of events resulting in plantarflexor weakness. Understanding the influence of physical activity on age-related differences in gait kinetics will determine if this shift is caused by low fitness rather than age, and if it avoidable by any means (e.g. a high level of physical activity), or if it is a natural and unavoidable consequence of normal, healthy aging.

If age-related differences in step length, trunk kinematics, and physical activity do not explain the age-related distal to proximal shift of joint kinetics in older adults, it may indicate age-related differences at the muscle level are contributing to the shift (e.g. muscle architecture and force production). Older adults, in general, have less gastrocnemius muscle mass (Lauretani et al., 2003; Narici et al., 2003), shorter muscle fascicle lengths (Narici et al., 2003), and smaller pennation angles (Morse et al., 2005; Narici et al., 2003) compared to younger adults. These structural differences are associated with age-related differences in maximum isometric force and slower contraction velocity in older adults (Thom et al., 2007). Therefore, older adults with shorter fascicle lengths and smaller pennation angles, compared to young adults, may also exhibit smaller muscle forces, and in turn, smaller moments and powers. Musculoskeletal modeling has been used previously to estimate muscle forces in older adults (Hasson & Caldwell, 2012; Schloemer et al., 2016; Thelen, 2003), but not with regard to explaining the age-related distal to proximal shift of kinetics in gait. Examining this phenomenon at a new level of complexity, could help inform whether this shift is due to negative and/or preventable changes in muscle architecture and resulting force production. The findings of these studies will serve as a starting point for future research aimed at determining if the age-related distal to proximal shift of joint kinetics is a normal and healthy part of aging, or if measures should be taken to prevent or delay this shift.

1.2 Problem Statement

Although age-related differences in walking mechanics have been well-documented, the mechanisms underlying these differences are unknown. Specifically,

there is little consensus about which biomechanical factors contribute to the age-related distal to proximal shift of joint moments and powers. Therefore, the purpose of this dissertation was to identify mechanisms contributing to the age-related kinetic differences in walking mechanics in an attempt to explain why older adults walk differently than young adults. This dissertation was designed to determine to what extent the age-related distal to proximal shift in joint kinetics can be explained by age-related differences in (i) step length and trunk kinematics, (ii) physical activity, and (iii) gastrocnemius muscle architecture and individual lower-extremity muscle forces.

1.3 Objectives and Hypotheses

This dissertation consisted of three studies focused on understanding why older adults walk differently than young adults. The study objectives and hypotheses are stated here. Figure 1 outlines the proposed mechanisms contributing to the age-related distal to proximal shift in joint kinetics.

Study 1

Objectives: 1.1)Determine the effects of walking with a step length similar to the step length of young adults on hip and ankle joint kinetics in highly physically fit older adults, and 1.2) determine the effects of walking with an upright trunk posture on hip and ankle joint kinetics in highly physically fit older adults.

Hypotheses: Walking with relatively shorter steps is associated with smaller ankle range of motion during stance and at push-off. Plantarflexion angle influences musculotendon dynamics such that smaller plantarflexion angles at pushoff contribute to a reduction in

force generating ability, and thus a reduction in plantarflexor moment. Additionally, ankle angle contributes to ankle kinetics because joint moments are calculated as a product of inertia and angular acceleration (a derivative of angular position). Therefore, **1.1) it is expected that when older adults walk with a step length similar to young adults, older adults will exhibit smaller hip joint kinetics and larger ankle kinetics compared to walking with a self-selected step length.** Additionally, the active torque-angle relationship of the hip extensor muscles indicate more torque can be generated at larger flexion angles, therefore, **1.2) it is expected that when maintaining an upright position during walking, older adults will display smaller hip extensor moments compared to walking with self-selected trunk lean.**

Study 2

Objectives: 2.1) Compare ankle and hip kinetics between non-runner older adults and endurance runner older adults to determine the effect of physical activity on the age-related distal to proximal shift of lower-extremity joint kinetics.

Hypothesis: Physically active older adults exhibit greater lower-extremity strength compared to non-active older adults, and since leg strength is associated with joint kinetics during walking (Silder et al., 2008), **2.1) it is expected that normally active older adults will display smaller ankle kinetics and larger hip kinetics compared to highly physically active older adults.**

Study 3

Objectives: 3.1) compare medial gastrocnemius fascicle length and pennation angle between young and older adults, and 3.2) predict individual lower-extremity muscle forces during walking in young and older adults to assess the age-related shift of joint mechanics at the muscle level. Due to the exploratory nature of this study and large number of variables, hypotheses were not established for specific muscles, but in general it is expected that due to the relationship between joint moment and muscle forces, older adults will display larger hip extensor muscle forces and smaller plantarflexor muscle forces.

CHAPTER 2

REVIEW OF LITERATURE

Aging is associated with a well-documented distal to proximal shift of joint moments and powers (Cofré et al., 2011; DeVita & Hortobagyi, 2000b; Monaco et al., 2009). In general, older adults exhibit smaller ankle joint moments and powers and larger hip joint moments and powers during walking compared to young adults. A number of mechanisms have been suggested as the source of the age-related distal to proximal shift in joint kinetics such as step length, plantarflexor weakness/dysfunction (DeVita & Hortobagyi, 2000b; Judge et al., 1996a), and even asymptomatic peripheral artery disease (Myers et al., 2016). However, the specific mechanisms for the age-related distal to proximal shift in joint kinetics remain unknown. Determining the source(s) of the shift may help to direct future research to determine if it is avoidable by any means, such as a high level of fitness, or if it is a natural and unavoidable consequence of normal, healthy aging. This literature review will examine potential explanations for the age-related distal to proximal shift of joint moments and powers in older adults. This chapter will focus initially on the basic kinematic variables of walking speed and step length, due to their associations with joint kinetics, and the remaining sections will discuss potential alternative explanations including joint kinematics, muscle strength, and muscle architecture.

Of note, the studies reviewed here are based on healthy, unimpaired, community-dwelling older adults (~60-65yrs or older), unless otherwise specified. It is necessary to limit this review to studies on "normal and healthy" older adults because differences between healthy and pathological gait deviate from differences between healthy young and

older adults. For example, older adults with chronic disease typically walk slower and with shorter step lengths compared to healthy older adults (McGibbon & Krebs, 2004; Myers et al., 2016).

2.1. Distal to Proximal Shift in Kinetics

During walking, older adults display differences in lower limb joint moments and powers compared to young adults (Cofré et al., 2011; DeVita & Hortobagyi, 2000b; Monaco et al., 2009). In general, these differences consist of smaller overall ankle plantarflexor moments and peak powers in older adults compared to young adults, and greater overall joint moments and peak powers at the hip and/or knee joints. DeVita and Hortobagyi (2000b) characterized this phenomenon as the "age-related mechanical plasticity of gait" and described a "distal to proximal shift" in joint kinetics, and postulated that it occurs in response to decreased physiological and biomechanical function of the plantarflexor muscles with age.

The age-related differences in ankle joint kinetics during walking appear consistently in the literature, with previous studies consistently reporting smaller ankle plantarflexor moment and/or power during late stance in older adults compared to young adults (Cofré et al., 2011; DeVita & Hortobagyi, 2000b; Judge et al., 1996a; Savelberg et al., 2007; Silder et al., 2008; Winter et al., 1990). Differences in hip and knee joint kinetics are thought to be compensatory in response to the change in ankle function, but have been reported less consistently than the differences in ankle joint kinetics. Previous studies have found smaller knee extensor impulse (DeVita & Hortobagyi, 2000b; Savelberg et al.,

2007), or no differences in knee kinetics between young and older adults (Boyer et al., 2012; Judge et al., 1996a). Similarly, at the hip, some studies have found greater peak hip extension moment or angular impulse (Boyer et al., 2012; Savelberg et al., 2007), greater peak hip flexion moment or angular impulse (DeVita & Hortobagyi, 2000b; Kerrigan et al., 1998) and positive power (DeVita & Hortobagyi, 2000b; Silder et al., 2008), or in some cases, no differences in hip joint kinetics (Judge et al., 1996a). A recent meta-analysis concluded that there is moderate support for the popular suggestion of age-related differences in ankle kinetics, specifically for smaller peak propulsive power in older adults compared to young adults, but no support for age-related differences in knee joint kinetics, and little support for age-related differences in hip joint kinetics (Boyer et al., 2017).

Although these age-related differences in gait kinetics have been well-documented, the mechanism(s) by which a decrease in ankle kinetics may occur with normal, healthy aging and may drive a shift of joint moments and/or powers elsewhere in the lower limb, is unknown. Two primary possibilities are extensively discussed in the literature (Sorenson & Flanagan, 2015): 1) aging results in altered joint moments and powers due to a decreased ability to generate lower-extremity joint moments and powers, which in turn affect kinematics, and 2) aging results in adaptations to kinematic strategies, resulting in kinetic adaptations. The following sections of this literature review will discuss potential contributors to the age-related distal to proximal shift in joint moments and powers.

2.2 Walking Speed

Many studies investigating the effects of age on gait mechanics have used self-selected walking speeds, which vary widely between young and older adults (Judge et al., 1996a; Kerrigan et al., 1998; Ko et al., 2009a; McGibbon & Krebs, 2004). Older adults typically choose a slower preferred walking speed than young adults (Judge et al., 1996a; Winter et al., 1990) and also select slower speeds when asked to walk at a self-selected "fast" speed (Ko et al., 2012).

Some of the in the differences noted in previous studies between young and older adults regarding hip and knee kinetics could be related to differences in walking speed, e.g. if both young and older adult groups walked at the same absolute speed, or at different but self-selected speeds, or some other comparison. If all kinetic variables have the same mathematical relationship with speed, then speed alone is unlikely to explain the distal to proximal shift because kinetics at one joint would scale similarly relative to the other joints. Conversely, if the hip and ankle joints respond differently to changes in walking speed, then speed is a more likely contributor to the age-related distal to proximal shift in joint kinetics. In support of walking speed as a potential contributor to the age-related distal to proximal shift in kinetics, some kinetic variables increase roughly linearly with speed, such as peak knee extension moment in late stance and peak knee flexion moment in early stance, while others increase roughly quadratically, such as peak hip flexion and extension moments, and peak positive ankle power (Lelas et al., 2003). However, when walking at the same speed as young adults, older adults continue to display smaller ankle joint moments and power compared to young adults (Cofré et al., 2011; Silder et al., 2008). Further, in a multiple regression analysis, Alcock et al. (2013) found that while age is a

strong predictor of peak plantarflexor moment, gait speed does not explain the age-related variance in plantarflexor moment, which suggests the age-related difference in plantarflexor moment is independent of walking speed.

To summarize, while self-selected walking speed may contribute to the magnitude of the age-related difference in ankle joint kinetics, it does not fully explain the age-related distal to proximal shift in kinetics, particularly when young and older adults both walk at the same absolute speed. There does not appear to be a common speed or a pair of unmatched speeds where young and older adults walk with similar joint kinetics.

2.3 Step Length

In addition to walking at a slower self-selected speed, older adults also take shorter steps compared to young adults. Step length typically increases with increasing walking speed, so it is not surprising that older adults walking at a self-selected speed use shorter step lengths compared to young adults. However, when older adults are asked to walk at the same speed as young adults, they typically still use a shorter step length compared to young adults (DeVita & Hortobagyi, 2000; Ko et al., 2009). Kinematically, shorter steps are associated with a smaller range of motion during stance and less plantarflexion at push off, compared to longer steps. Since joint angular position is indirectly included in the calculation of joint moment, these shorter step lengths may contribute to smaller plantarflexor moment and power observed in older adults. Indeed, ankle positive power correlates strongly with step length in both young and older adults (Judge et al., 1996). Further, Manipulation of step length in young adults resulted in changes to kinetics such

that walking with a shorter step length decreased ankle moment, power, and impulse, and increased hip moment and power (Umberger & Martin, 2007). In effect, by walking with shorter steps, young adults showed a shift of joint moments and power similar to the distal to proximal shift that is theorized to occur with aging. Based on these findings, one might postulate that if older adults were to walk with an increased step length (similar to the step length of young adults) at a standard speed, joint moments might shift to resemble those of young adults. However, the effect of systematic adjustments to step length on gait kinetics in older adults has not been investigated to date.

There is also evidence indicating that shorter step lengths are not the sole contributor to smaller plantarflexor moments with age. Some studies have reported differences in hip and ankle moments and powers between young and older adults even when walking with similar step lengths. For example, Silder et al. (2008) compared net positive hip and ankle work when young and older adults walked at similar self-selected speeds and step lengths. They found greater net positive work at the hip and smaller net positive work at the ankle in older compared to young adults. Anderson and Madigan (2014a) compared joint moments when young and older adults walked at slow (1.1 m/s) and fast (1.5 m/s) speeds. Step lengths were similar between groups at both speeds. During fast walking, they found a smaller peak plantarflexor moment and greater peak hip extensor moment in older compared to young adults, but during slow walking, found no differences in peak plantarflexor or hip extensor moment between young and older adults. The findings of Silder (2008) and Anderson and Madigan (2014) show that even when walking with similar self-selected step lengths, young and older adults exhibited differences in

ankle and hip joint kinetics, providing evidence against step length as the sole contributor to the age-related distal to proximal shift in joint kinetics.

In the studies mentioned above, older adults either circumstantially walked with step lengths similar to young adults (Silder et al., 2008) or walked only with a prescribed step length (Anderson & Madigan, 2014). The effect of walking with a longer vs. shorter step lengths on joint kinetics in older adults walking at a standard speed has not been directly investigated. If older adults are able to walk with joint kinetics similar to young adults simply by increasing step length, it would suggest that kinematics play a large contributing role in the age-related distal to proximal shift of joint moments and powers, yet, for some reason, older adults preferentially walk with shorter step lengths and, in effect, characteristically altered joint kinetics. Identifying the effect of step length on joint kinetics in older adults would inform future research aiming to determine the cause of the age-related preference towards shorter step lengths, such as increased stability during walking.

2.4 Sensory and Cognitive Considerations

Older adults often report peripheral neurological deficits such as decreased vibratory sensation and proprioception in their feet (Mold et al., 2004). Peripheral neuropathy is associated with weakness, numbness, and pain in the hands and feet, and is most commonly observed as a secondary health condition in individuals with various chronic diseases, such as diabetes, but idiopathic peripheral neuropathy has also been observed in active, healthy older adults (Mold et al., 2004). A reduced ability to sense the

position of the foot relative to the body/ground or feel the ground beneath the foot (particularly during push-off) could compromise range of motion at the ankle and force production at push-off, leading to differences in gait mechanics between young and older adults. For example, experimentally induced plantar insensitivity results in decreased plantar pressures (Taylor et al., 2004), and individuals with peripheral neuropathy display smaller anterior-posterior ground reactions forces compared to individuals without peripheral neuropathy (Mueller et al., 1994). Further, individuals with peripheral neuropathy walk slower and with shorter step lengths compared to individuals without peripheral neuropathy (Mold et al., 2004; Mueller et al., 1994). In addition, diabetic individuals with peripheral neuropathy exhibit a smaller peak ankle moment and smaller peak ankle power compared to nondiabetic individuals. Thus, healthy older adults with undiagnosed idiopathic peripheral neuropathy may also exhibit smaller ankle joint kinetics compared to people without peripheral neuropathy, contributing to the well-documented observations of an age-related distal to proximal shift in joint kinetics.

In addition to sensory decrements, older adults commonly report a fear of falling (Arfken et al., 1994). Older adults with a fear of falling may engage compensatory strategies to increase stability, which would contribute to spatiotemporal differences, and thus result in kinetic differences between individuals with and without a fear of falling. For example, fear of falling has been associated with slower walking speed, shorter step length, and a greater step width (Chamberlin et al., 2005). These spatiotemporal variables are thought to increase stability through prolonging the time spent in stance and double support, and may also affect lower-extremity joint kinetics.

Peripheral neuropathy and fear of falling may alter spatiotemporal parameters in a manner consistent with the effects of aging. However, as discussed in the previous sections, it is unlikely that walking speed and step length are the sole contributors to the age-related distal to proximal shift of joint kinetics. Therefore, any effects of peripheral neuropathy and fear of falling on otherwise healthy, active adults are probably minimal with regard to the age-related distal to proximal shift of kinetics.

2.5 Muscle Strength

Older adults experience a well-documented reduction in maximal strength. Cross-sectional studies consistently report smaller peak isometric and concentric forces in older adults compared to young adults (Lindle, 1997; Mitchell et al., 2012). This age-related decrease in muscle strength could contribute to the age-related distal to proximal shift in kinetics through a smaller capacity to produce muscle forces, resulting in smaller moment and power generation in older adults compared to young adults.

Muscle strength is strongly correlated to parameters of gait kinetics. For example, maximum isometric plantarflexor torque correlates positively with peak plantarflexor power during walking (Silder et al., 2008). Similarly, musculoskeletal modeling suggests that decreased plantarflexor strength may play a role in the age-related shift of joint kinetics. For example, a 30% simulated reduction in maximum isometric plantarflexor strength resulted in increased hip extensor mechanical work, indicating a compensatory relationship between ankle and hip joint kinetics (Goldberg & Neptune, 2007). The impact of plantarflexor weakness on normal walking ability is further supported by findings that a

greater percentage of maximal muscle force is required from the plantarflexor muscles during walking than is required from other muscles of the lower-extremity (Kulmala et al., 2016). Thus, walking gait mechanics appear to be particularly sensitive to the level of available strength in the plantarflexors, and weakness in this muscle group can at least in theory contribute to decreased ankle joint moment and power generation. In fact, several authors have speculated that age-related differences in plantarflexor moment and power during gait are associated with general plantarflexor weakness in older adults (Cofré et al., 2011; DeVita & Hortobagyi, 2000b; Judge et al., 1996a).

Age-related differences in plantarflexor strength (Bemben et al., 1991; Christ et al., 1992) are comparable to reported age-related differences in knee extensor strength (Lanza et al., 2003; Macaluso et al., 2002) and hip extensor strength (Dean et al., 2004). Since the hip, knee, and ankle joint display similar differences in strength between young and old adults (as opposed to the ankle displaying a greater age-related difference compared to the hip and/or knee), it is unclear why the ankle kinetics, specifically, are so consistently affected (Figure 2), thought it may be related to structural changes at the muscle level. Further, there is evidence that despite the well-documented reduction in plantarflexor moment and power, older adults retain a substantial but underutilized capacity for moment and power generation at the ankle (Franz, 2016). That is, during walking, older adults use a fraction of their maximal plantarflexor capacity, and walking mechanically like a young adult would not seem to require supra-maximal muscle forces and powers. Therefore, decreased strength and a reduced capacity for ankle power production does not satisfactorily explain the age-related differences in gait.

2.6 Joint and Segment Kinematics

Similar to age-related differences in lower-extremity joint kinetics, kinematic differences at the ankle are also well-supported in the literature, but differences at the knee and hip are somewhat less consistent. Older adults walk with a smaller ankle range of motion (DeVita & Hortobagyi, 2000b) and a smaller peak plantarflexion angle compared to young adults (Boyer et al., 2017; Cofré et al., 2011; DeVita & Hortobagyi, 2000b; Kerrigan et al., 1998; Monaco et al., 2009; Silder et al., 2008), maintaining a more neutral ankle position throughout stance.

At the hip, older adults exhibit a greater range of motion (DeVita & Hortobagyi, 2000b; Kerrigan et al., 1998; Silder et al., 2008), greater peak flexion (Anderson & Madigan, 2014a; Cofré et al., 2011), and less peak extension (Anderson & Madigan, 2014a; Kerrigan et al., 1998; Riley et al., 2001), compared to young adults, suggesting older adults walk with a more bent over or trunk leaning posture than young adults.. Considering that the hip extensor muscles typically can produce more torque at larger flexion angles (Anderson et al., 2007), increased hip flexion during stance due to forward trunk lean may result in an increased hip extensor moment and propulsive power, yet to date, there have been no systematic investigations of the effect of trunk lean on the age-related shift of joint kinetics in older adults. Older adults exhibit larger hip flexion angles even when controlling for speed or step length (Anderson & Madigan, 2014; Boyer et al., 2017; Cofre et al., 2011), which indicates that age-related differences in hip and/or trunk kinematics, independent of age-related differences in speed and step length, may be a primary contributor to the age-related distal to proximal shift in joint moments and powers.

2.7 Muscle Architecture and Mechanical Properties

Muscle architecture refers to the arrangement of a muscle's fascicles, with specific regard to their length and orientation, and is a primary determinant of various physiological aspects of muscle force production (Gans & De Vree, 1987). For example, the maximum shortening velocity of a muscle is proportional to the number of sarcomeres arranged in-series, and thus, is proportional to the length of a muscle fascicle (Bodine et al., 1982; Burkholder et al., 1994). Muscles with relatively long fascicles can produce forces over a larger operating length at high shortening speeds due to the simultaneous contraction of serially arranged sarcomeres (Spector et al., 1980). Additionally, a longer fascicle can generate greater force than a shorter fascicle because for a given whole-fascicle shortening velocity, the increased number of sarcomeres allows each sarcomere to operate at a slower speed at which more force can be produced.

Similarly, the number of in-parallel fascicles substantially contributes to a muscle's physiological cross sectional area (PCSA), which is roughly proportional to maximum isometric force (Haxton, 1944; Lieber & Friden, 2000). Theoretically, PCSA represents the cross-sectional area of a muscle that is perpendicular to all of its fascicles (Haxton, 1944), and is calculated as a ratio of muscle fascicle volume to muscle fascicle length along the axis of pennation. The consideration of pennation angle in this calculation is particularly important for multi-pennate muscles such as the gastrocnemius because when fascicles are oriented at an angle relative to the force-generating axis, not all of the force is transmitted to the tendon (Figure 3) (Alexander & Vernon, 1975). Although a smaller pennation angle increases the proportion of force directed along the tendon, a greater pennation angle is thought to improve force generation through: 1) allowing for a larger

PCSA (i.e. more fascicles packed in along the tendon, acting in parallel), 2) allowing fascicles to operate closer to the length at which maximum force can be produced by shortening less for a given tendon excursion (due to fascicle rotation of pennate muscle), and 3) reducing fascicle velocity for a given whole-muscle shortening velocity (Blazevich, 2006).

In summary, muscles with relatively short fascicles and relatively small pennation angles will produce less force and the production of force will be more sensitive to shortening velocity than a muscle with the same volume but longer fascicles or a greater pennation angle (Figure 4). These functional differences are important because fascicle length and pennation angle can change with atrophy of the muscle (Narici & Cerretelli, 1998). Considering that muscle atrophy is a common occurrence in older adults (Evans & Lexell, 1995; Mitchell et al., 2012; Roubenoff & Hughes, 2000), it follows that architectural differences have also been observed in older adults compared to young adults. For example, older adults exhibited smaller pennation angles in the medial gastrocnemius, lateral gastrocnemius, and soleus muscles, and shorter fascicle length in the medial gastrocnemius, compared to young adults (Morse et al., 2005). However, effects of age on fascicle length and pennation angle are not consistently reported. Morse et al. (2005) found no age-related differences in fascicle lengths of the lateral gastrocnemius or soleus muscles. Similarly, Randhawa and Wakeling (2013) found no age-related differences in lateral or medial gastrocnemius fascicle length, but found smaller pennation angles for both muscles in older adults.

A point of contention in the literature is whether or not age-related differences in muscle architecture are truly due to "aging" or, rather, are a result of the sedentary lifestyle

and decreased physical activity often associated with aging. In support of age (as opposed to physical activity) as a contributing factor to changes in muscle architecture, Narici et al (2003) compared the fascicle lengths and pennation angles of the medial gastrocnemius in young and older adults who were matched for height, mass, and physical activity, and still found shorter fascicle lengths and smaller pennation angles in older adults. The authors suggest that the effect of aging contributed to the differences in muscle architecture between young and older adults, rather than disuse. In contrast, Karamanidis and Arampatzis (2005) found no differences in medial gastrocnemius pennation angle or fascicle length between young and older endurance runners, but they also found no age-related differences in non-active individuals. The influence of fitness on muscle architecture in older adults is therefore unclear.

Muscle architecture is closely related to the maximal isometric strength and maximal shortening velocity of a muscle. Of particular interest are the force-length and force velocity relationships that define the contractile mechanics of muscle in the typical Hill-type model. Despite the structural changes to muscle with age, the force-length relationship of older adults is generally observed to be similar to that of young adults (Van Schaik et al., 1994; Winegard et al., 1997). For example, although the plantarflexor muscles of young adults are stronger and elicit greater isometric twitch torque compared to older adults, the angle at which maximum isometric torque can be produced is the same for young and older adults (Winegard et al., 1997)(Figure 5).

In summary, it is likely that age-related differences in muscle architecture play a contributing role in the age-related distal to proximal shift of joint kinetics through changes

in muscle force-length velocity properties and associated changes in muscle force production.

2.8 Physical Activity

Frequent physical activity contributes to the maintenance of muscle strength, muscle mass, and structure with age (Sandler et al., 1991; Visser et al., 2002), but many older adults report declining levels of physical activity and substantial mobility challenges (Hallal et al., 2012; Milanović et al., 2013). Since strength and muscle structure probably contribute to the age-related distal to proximal shift (as covered in previous sections), it is possible that the age-related distal to proximal shift is associated with physical activity level in older adults. For example, the age-related distal to proximal shift was partially attenuated in older adults who engaged in a high volume of daily walking (at least 7500 steps/day, but did no running or jumping/impact activities) compared to young adults with similar physical activity level (Boyer et al., 2012). However, the age-related distal to proximal shift was still present in older adults who engaged in running activities at least twice a week for two years (Savelberg et al., 2007) and in older adults who performed at least 30 minutes of moderate activity at least twice a week for a year (Buddhadev & Martin, 2016). Thus, the effect of physical activity on age-related kinetic differences is unclear, but may at least partially mediate the effects of aging on joint kinetics.

In the studies mentioned above, the definition of 'physically active' varies. Thus, conclusions about the effect of physical activity on gait kinetics in older adults are ambiguous. One possible way of accounting for the effect of physical activity on gait

kinetics in older adults is by recruiting highly active older adults, such as master's level endurance runners, who theoretically represent a model of successful aging (Bortz IV & Bortz, 1996; Hawkins et al., 2003). Comparisons of highly physically active older adults and healthy, inactive (or lightly active) older adults can help to clarify the effect of physical activity with age. For example, if highly physically active and inactive older adults do not exhibit differences in gait kinetics, it would indicate that physical activity does not reverse or reduce the age-related distal to proximal shift, and may suggest that this shift is a natural and unavoidable part of aging. Further, investigating age-related differences in this population (by comparing to highly active young adults) can account for the potential confounding relationship of physical activity and allow for interpretation of differences based on the effect of age.

2.9 Effects of Modeling/Data Processing

In addition to potential mechanical or physiological contributions to the age-related distal to proximal shift, data processing techniques may also contribute to age-related differences in joint moments and powers. Due to variance in data processing techniques or experimental methods, these data processing issues should be considered when interpreting age-related differences. It is generally accepted that a certain amount of 'noise' is present in marker position data due to skin motion artifact and marker placement error. Skin motion artifact refers to the vibration or movement of the skin relative to movement of the bony landmark the marker is intended to identify. Typically, errors due to skin motion artifact are observed during high impact movements such as running or jumping (Kristianslund et al., 2012; Van den Bogert & De Koning, 1996), and in regions

of the body with relatively greater adipose tissue, such as the thigh compared to the shank (Angeloni et al., 1994). Marker position data are low-pass filtered to improve the signal to noise ratio, or 'smooth' the data. Cutoff frequencies for kinematic data are typically low (6-10 Hz) in order to eliminate the most noise with least attenuation of the signal. It is known that the optimal frequency content (based on Winter's residual analysis (Winter, 2009) of various segments are different (Angeloni et al., 1994), with some suggesting that each marker or segment be filtered at its optimal cutoff frequency to minimize the amount of signal attenuation.

While unlikely, techniques in accounting for this high-frequency noise during data processing may contribute to the observed distal to proximal shift of kinetics. Differences in body composition and skin elasticity between young and older adults may result in age-related differences to the optimal cutoff frequency for each marker or segment. In general, gait studies apply one cutoff frequency to all kinematic data and participant groups. If the optimal cutoff frequency content differs between groups, filtering at the same cutoff may introduce errors in kinematic data which are then amplified when differentiating to estimate segment accelerations. For example, errors in joint moments due to erroneous segment accelerations in the inverse dynamic equations are largest at the hip (Kristianslund et al., 2012; Van den Bogert & De Koning, 1996) because they depend on calculations from distal segments (Davis et al., 1991). Therefore, the larger hip joint moments and powers observed in older adults may be at least partially due to cumulative artifact error from skin motion and the resulting segment accelerations used in inverse dynamics. However, errors on this magnitude are unlikely, particularly for flexion/extension moments of the sagittal plane which has been shown to have the least amount of error due to skin motion artifact

compared to the frontal and transverse planes (Leardini et al., 2005; Reinschmidt et al., 1997)

2.10 A Note about Cross-Sectional Studies

All of the studies reviewed thus far were cross-sectional. Therefore, based on these studies alone, the assumption that the age-related differences in kinetics and kinematics are changes that occur with aging is indeed an assumption. Longitudinal studies on aging are rare due to the length of follow up time necessary to make conclusions about the effect of 'aging'. However, exercise intervention studies often record data at baseline and at least one other time point, providing a small window of longitudinal data. Beijersbergen et al. (2013) conducted a review of studies on strength training in older adults and while the older adults in the reviewed studies often increased strength, the authors found limited support for the association between strength gains and changes in gait kinetics or kinematics. Therefore, older adults do not seem to consistently utilize these gains in strength to modify joint kinetics, suggesting that perhaps the mechanisms accounting for the age-related changes in kinetics are due to parameters associated with muscle force production, such as muscle architecture and its effect on mechanical properties.

2.11 Summary

The (cross-sectional) effect of age on gait mechanics has been extensively documented. Older adults exhibit smaller ankle moment and power compared to young adults, and in some cases also exhibit greater moments and/or power generation at the hip

and/or knee. The mechanisms by which this distal to proximal shift occurs are unknown. However, differences in walking speed do not appear to be a major factor, and previous studies have shown that step length affects ankle kinetics, but the effect of walking with an increased step length while controlling speed has not been directly investigated in older adults. It is unknown if there are particular step lengths at which young and older adults have similar joint kinetics. Similarly, there have been no systematic investigations regarding the effect of trunk position on joint kinetics in older adults. If step length or trunk position contribute to the shift of joint moments and powers with age, future studies can be designed to determine the reasons for the age-related preference towards shorter step lengths and/or increased trunk flexion and smaller ankle plantarflexor moment and power in older adults, such as increased stability or minimizing metabolic cost.

Beyond step length and trunk position, the effect of habitual physical activity on gait kinetics in older adults is unclear. While it is thought that regular physical activity helps to maintain muscle strength and mobility in older adults, physical activity levels decline with age. Therefore, it is possible that physical activity level, independent of age, is at least partially contributing to the age-related distal to proximal shift in gait kinetics. If highly physically active older adults and normally physically active older adults do not exhibit differences in gait kinetics, it may indicate that physical activity does not reverse or reduce the age-related distal to proximal shift, and may further suggest that this shift is a natural and unavoidable part of aging.

Other possible contributors to age-related differences in walking mechanics are muscle architecture and mechanical properties. While it is known that muscle architecture differs between young and older adults, the distal to proximal shift has not been

investigated at the muscle level. Understanding how muscles contribute to the age-related differences in joint kinetics will inform future research seeking to determine if this age-related shift is due to preventable or modifiable features of muscle architecture and muscle force production. In summary, the most promising biomechanical explanations for the source of the age-related distal to proximal shift in joint kinetics are step length or trunk kinematics, history of exercise and current physical fitness, and lower limb muscle architecture and muscle forces. Specific unanswered questions that could clarify how these factors mechanistically affect joint torques during older adult gait are:

1. Can older adults walk with joint kinetics similar to young adults by walking with a longer step length, or more upright posture, similar to young adults?

2. Does physical activity reduce or reverse the distal to proximal shift in older adults?

3. Do features of gastrocnemius muscle architecture differ between young and older adults? Relatedly, is the age-related distal to proximal shift evident at the muscle force level, and if so, are age-related differences apparent in all muscles acting within a muscle group, or just specific muscles?
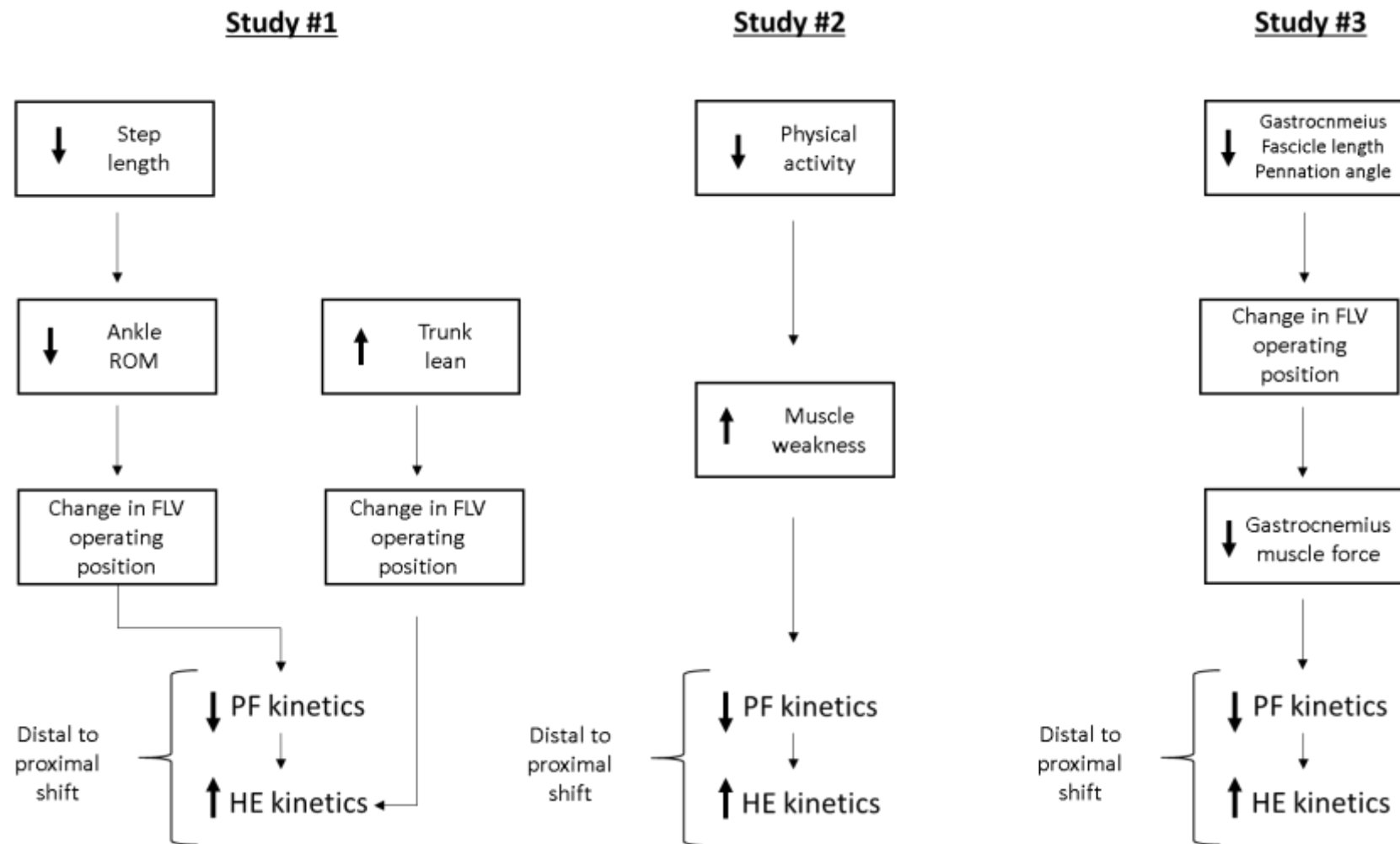
Figure 2.1. Proposed mechanisms of age-related distal to proximal shift of joint kinetics. ROM (range of motion), PF (plantarflexor), HE (hip extensor), FLV (force-length-velocity).
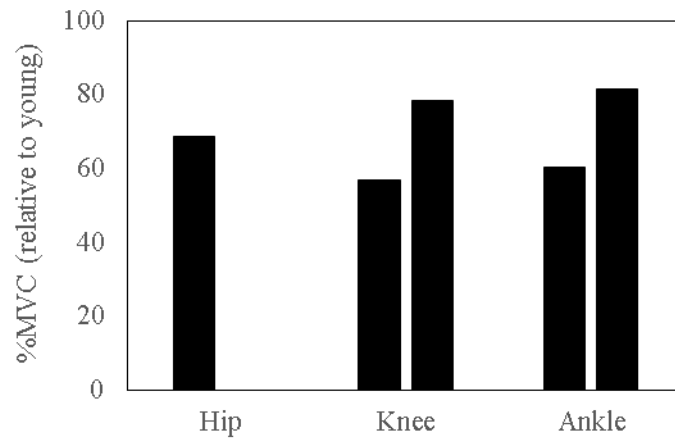
Figure 2.2. Maximal voluntary contractions (MVC) during isometric extension or plantarflexion tasks in older adults expressed as a percentage of young adult MVC for the hip (Dean et al., 2004), knee (Macaluso et al., 2002; Lanza et al., 2003) and ankle (Christ et al., 1992; Bemben et al. 1991).
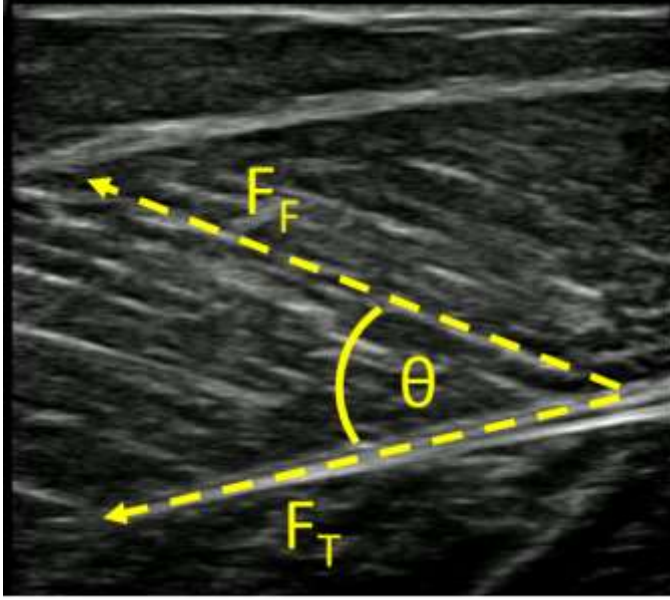
Figure 2.3. When fascicles are oriented at an angle (θ) the force transmitted to the tendon (F_T) is a fraction of the force produced by the fascicles (F_F). $F_T = F_F \theta$.
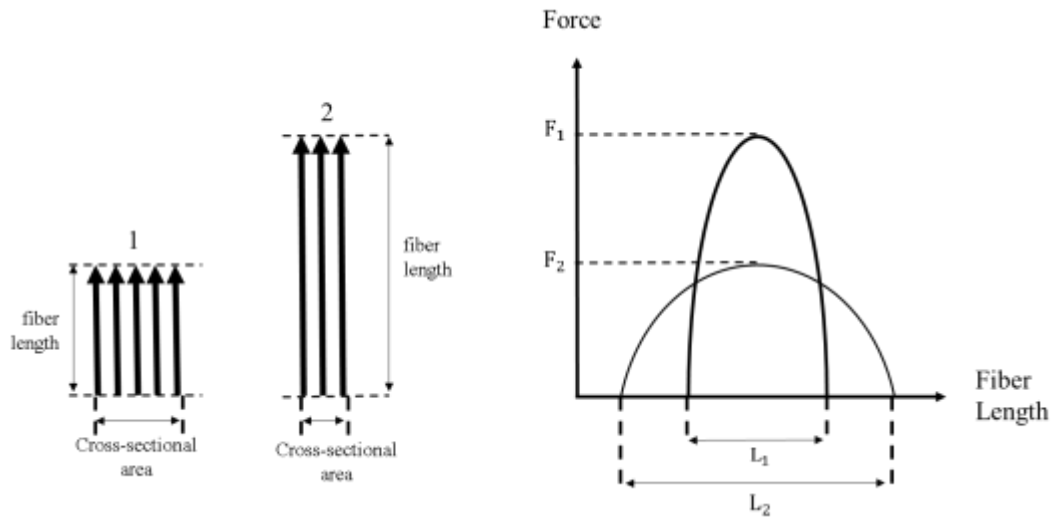
Figure 2.4. Adapted from Nigg & Herzog *(2003)*. Schematic force-length relationship illustrating two muscles with different fascicle lengths and cross-sectional areas, but equal volumes.
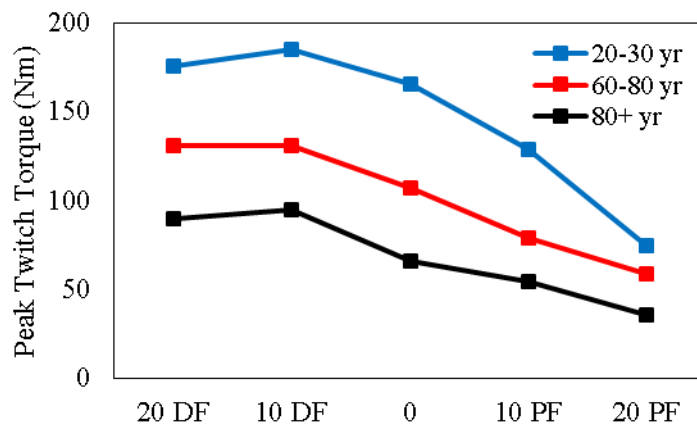
Figure 2.5. Adapted from Winegard et al. (1997). MVC values for 15 males in three age groups at 5 ankle positions (DF=dorsiflexion, PF=plantarflexion). All age groups displayed the highest MVC at the same ankle position (10 degrees of dorsiflexion).

CHAPTER 3

STEP LENGTH AND TRUNK FLEXION ANGLE DO NOT REVERSE THE AGE-RELATED DISTAL TO PROXIMAL SHIFT OF JOINT KINETICS IN OLDER ADULTS

3.1 Introduction

Older adults typically take shorter steps compared to young adults, even when walking at the same speed as young adults (DeVita & Hortobagyi, 2000b; Ko et al., 2009b; Winter et al., 1990). These characteristic shorter steps are posited as a potential mechanism to explain the larger hip kinetics and smaller ankle kinetics often observed in older vs. young adults (i.e. distal to proximal shift in kinetics). For example, older adults who walk with shorter step lengths also have smaller ankle range of motion during stance phase of walking, and smaller plantarflexion angle at pushoff (Judge et al., 1996a; Murray et al., 1969), potentially limiting the amount of plantarflexor force that can be generated via the muscle force-length relationship (Arnold & Delp, 2011). Consequently, the characteristic shorter step length of older adults may contribute to the age-related shift in kinetics via a reduction in force-generating capacity about the ankle due to joint kinematics. In support of this speculation, step length has been identified as a strong correlate of ankle joint power in older adults (Judge et al., 1996), suggesting that modifications to step length may, in turn, modify ankle joint kinetics. Modifying step length results in changes to ankle kinetics in young adults, such that shorter step length is associated with decreased ankle moments, powers, and impulses (Allet et al., 2011; Umberger & Martin, 2007). However, there have been no systematic investigations of step length on gait kinetics in older adults, therefore

it is unknown if older adults walking with a step length similar to young adults will exhibit an increased ankle moment.

The age-related differences in lower-extremity joint moments and powers (DeVita & Hortobagyi, 2000b) may be related to the substantial mobility challenges that occur with age, such as slower walking speed (DeVita & Hortobagyi, 2000b; Riley et al., 2001) and reduced metabolic efficiency (Huang et al., 2015), compared to young adults. One of the hallmarks of the age-related distal to proximal shift is the notable reduction in plantarflexor power during pushoff (Franz, 2016). The hip is largely thought to compensate for decreased plantarflexor power at pushoff by generating a larger peak hip extensor moment and power in early stance, larger hip moment impulse and positive hip work throughout stance, and larger hip flexor moment in late stance, to propel the body forward and pull the leg into swing, respectively (DeVita & Hortobagyi, 2000b; McGibbon, 2003). However, it is possible that age-related differences in hip kinetics are not solely a result of compensation for lack of ankle power, but may be directly influenced by trunk kinematics. Specifically, older adults walk with greater hip flexion throughout the gait cycle (DeVita & Hortobagyi, 2000b; Judge et al., 1996a; Kerrigan et al., 1998), suggested to be a result of increased forward trunk lean (DeVita & Hortobagyi, 2000b) and/or hip flexion contractures (Kerrigan et al., 1998; Silder et al., 2008). Considering that the hip extensor muscles typically can produce more torque at larger flexion angles (Anderson et al., 2007), increased hip flexion during stance due to forward trunk lean may result in an increased hip extensor moment and propulsive power. Thus, it is possible that age-related differences in hip kinetics are not solely a result of compensation at the hip joint due to reduced ankle

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plantarflexor moment and power generation, rather, these differences may be may be partially influenced by trunk kinematics.

Age-related declines in physical activity have also been suggested as a potential contributor to the age-related differences in joint kinetics. Older adults who maintained a high level of activity (≥ 7500 steps/day) displayed a small reversal of the characteristic distal to proximal shift observed in older adults (Boyer et al., 2012). However, there have been no studies investigating this shift in highly physically fit older adults (e.g. master's athletes), as a means of accounting for the potential confounding relationship of physical fitness. To our knowledge, there have been no systematic investigations of step length and trunk kinematics in relation to lower-extremity joint kinetics in older adults of any fitness level or with a high level of fitness, specifically. Addressing this gap in the literature will help elucidate mechanisms responsible for the age-related shift in joint moments and powers. For example, can the distal to proximal shift in joint kinetics be reversed, in whole or in part, when older adults walk with longer step lengths and a more upright trunk, characteristic of young adult gait? Moreover, characterizing the influence of trunk kinematics on age-related differences in hip kinetics may help clarify the compensatory role of the hip joint in the distal to proximal redistribution of lower extremity kinetics observed in older adults.

The objective of this study was two-fold: 1) to determine the effects of manipulating step length in older adults such that they walk with a step length similar to the step length of young adults, on hip and ankle joint kinetics in highly physically fit older adults, and 2) independent of objective one, determine the effects of walking with an upright trunk posture on hip and ankle joint kinetics in highly physically fit older adults. Due to changes

in joint kinematics that occur when walking with shorter or longer step lengths, we hypothesized that when the step length of older adults was modified to represent a step length similar to young adults, older adults would exhibit smaller hip joint kinetics and larger ankle kinetics compared to walking with a self-selected step length. Additionally, the active torque-angle relationship of the hip extensor muscles indicate more torque can be generated at larger flexion angles. Therefore, we also hypothesize that when maintaining an upright position during walking, older adults would display smaller hip extensor moments compared to walking with self-selected trunk lean.

3.2 Methods

Participants

14 young adult men and 14 older adult men were recruited and tested for this study. However, one young and one older adult were excluded from analysis due to poor motion tracking data, and one older adult was excluded due to equipment malfunctions that prevented data collection for one of the walking conditions. Therefore, participants were 12 older adult men (67 ± 5 yrs, 1.79 ± 0.07 m, 77.3 ± 13.7 kg), and 13 young adult men (21 ± 3 yrs, 1.80 ± 0.05 m, 70.3 ± 5.0 kg). The minimum detectable difference with error rates of 5% for false positive and 20% for false negative was $dz = 1.0$, which is similar to or greater than the effects in other studies on modified step lengths (Allet et al., 2011). Prior to enrolling in this study, individuals were screened and excluded if they reported previous injuries to the legs or back that required medical attention and/or chronic medical conditions that have been associated with gait changes such as diabetes and peripheral artery disease (Myers et al., 2016). Both young and older adults were highly physically

active, and self-reported running at least 20 miles/week and training for at least one race of 10 km distance or longer in the past year. The older adults each self-reported that they had been running for at least the past seven years, with most reporting 20-30 years or more of training. Age-related declines in physical activity may partially contribute to age-related differences in kinetics, therefore, highly physical active participants (i.e. endurance runners) were chosen in order to limit the potentially confounding nature of physical activity with age (e.g. Boyer et al. 2012). All participants provided written informed consent to procedures approved by the University of Maryland Institutional Review Board.

Experimental Setup

Participants wore shorts and their own athletic shoes. Positions of 44 retroreflective markers on the pelvis and lower limbs (Fig. 1) were captured at 200 Hz using a 13-camera motion capture system (Vicon, Oxford, UK). Force plate data were captured at 1000 Hz using 10 six degree of freedom piezoelectric force plates (Kistler, Switzerland) arranged in a single row located within a raised platform.

Study Design

All young adult participants were enrolled and tested prior to enrolling any older adult participants. Young adults completed five walking trials at a standard speed of 1.3 m/s with a self-selected step length. The preferred step lengths measured in the young adult group were then used to set the stride length conditions for the older adult group. Step length for each older adult was selected from a young adult of similar height. The average step length for younger adults walking at 1.3 m/s was 0.73 ± 0.03 m.

Older adults completed five walking trials in each of three conditions: 1) standard speed of 1.3 m/s with a self-selected step length (1.3-SS), 2) standard speed of 1.3 m/s while matching the prescribed step length from the young adult group (1.3-SL), and 3) standard speed of 1.3 m/s while keeping their trunk upright (1.3-Trunk). The order of the 1.3-SL and 1.3-Trunk conditions was counterbalanced between subjects. After each trial participants were given verbal feedback about their speed, and, for the 1.3-Trunk condition, participants were reminded to walk while keeping their trunk as upright as possible. Speed was monitored using an infrared timing gate system and trials in which the average over the force plates was not within $\pm 3\%$ of 1.3 m/s were excluded. Step length was constrained indirectly by controlling for both walking speed and step frequency via an auditory metronome. Since walking speed is equal to the product of cadence and step length, holding speed constant and adjusting cadence resulted in changes to step length.

Data Processing

Five clean footstrikes isolated on a single force plate were selected for analysis in each condition. Marker position and ground reaction force data were exported to Visual3D (C-Motion, Germantown, MD, USA) and smoothed using a 4th-order dual-pass Butterworth filter with cutoff frequencies of 6 Hz and 45 Hz, respectively. A linked-segment model was created for each participant from marker positions during a standing calibration trial. The trunk segment was defined distally with the left and right acromion process markers, and proximally at the L5S1 joint. Trunk angle was calculated with respect to the lab coordinate system and normalized to static standing. Joint angles were calculated using 6DOF pose estimation and a Cardan Xyz rotation sequence (Wu & Cavanagh, 1995). Iterative Newton-Euler inverse dynamics within Visual3D were used to calculate joint

powers and moments (Selbie et al., 2014). Joint moment impulse was calculated as the time integral of the moment curve, and joint work was calculated as the time integral of the power curve. Joint moment impulse and joint work were calculated for the stance phase, defined as heel strike to toe-off (~0 – 60% of the gait cycle).

Statistical Analysis

Outcome variables were peak sagittal plane hip extensor and flexor moments and impulse, peak ankle plantarflexor moment and impulse, and peak hip and ankle positive powers and joint work. Variables were determined from each trial and averaged over trials to produce a representative value for each participant. All variables were scaled by participant mass. To test for the effect of stride length, a one-tailed paired *t*-test was conducted to determine differences in outcome variables between 1.3-SS and 1.3-SL. To test for the effect of trunk position, a one-tailed paired *t*-test was conducted to determine differences in outcome variables between 1.3-SS and 1.3-Trunk. One-tailed *t*-tests were used due to the directional nature of the hypotheses. Bonferroni correction was used to adjust for multiple comparisons, resulting in critical $p = 0.005$. Effect sizes (Δ) were calculated, based on Glass' Delta, as the difference in means between conditions divided by the standard deviation of young adults in the 1.3-SS condition. The standard deviation of young adults were used in this calculation in order to interpret effect sizes in reference to between-subjects variance that is presumably normal and healthy. Young and older adults did not display large differences in variability, therefore the while effect sizes reported here would be similar to more traditional effect size calculations (Cohen's *dz*), effect sizes calculated based on Glass' Delta allow for easier interpretation of meaningfulness.

3.3 Results

Average step lengths for the older adult group in each condition were: 1.3-SS (0.72 ± 0.04 m), 1.3-SL (0.73 ± 0.03 m), and 1.3-Trunk (0.73± 0.04 m) and average walking speeds were: 1.3-SS (1.30 ± 0.02 m/s), 1.3-SL (1.30 ± 0.02 m/s), and 1.3-Trunk (1.31 ± 0.01 m/s). The average step length and walking speed for the young adult group in the 1.3-SS condition were 0.73 ± 0.03 m and 1.31 ± 0.02 m/s, respectively.

Differences in joint kinetics between young and older adults walking in the 1.3-SS condition were consistent with the distal to proximal shift in joint moments and powers reported elsewhere (e.g. DeVita & Hortobagyi, 2000): older adults displayed 46 and 55%% smaller peak hip flexor moment and impulse, respectively, 33% greater hip extensor moment impulse during stance, and 40% more positive work at the hip during stance, compared to young adults (Fig. 2 and 3). Additionally, older adults exhibited a 16% smaller ankle plantarflexor moment impulse and generated 13% less positive work at the ankle compared to young adults. The magnitude of peak ankle moment and power appeared similar between young and older adults, which may be due to the high physical activity level of both groups (Fig. 2 and 3).

Effect of Step Length on Joint Kinetics in Older Adults

Older adults in this sample of highly active individuals self-selected step lengths similar to young adults when walking in the 1.3-SS condition. As a result, the prescribed step length in the 1.3-SL condition was not different on average from the 1.3-SS condition,

and there were no differences in outcome variables between 1.3-SS and 1.3-SL conditions (all $p > 0.005$; Table 1 and 2, Fig. 2 and 3).

Effect of Trunk Kinematics on Joint Kinetics in Older Adults

In the 1.3-Trunk condition older adults walked with ~3 degrees less trunk flexion, compared to the 1.3-SS condition (Fig. 4). However, there were no effects of trunk position on any outcome variable (all $p > 0.005$, Table 1). The magnitude and shape of the moment and power curves were similar between conditions (Fig. 2 and Fig. 3).

3.4 Discussion

The purpose of this study was to 1) determine the effect of stride length on lower-extremity joint moments and powers in older adults, and 2) determine the effect of walking with an upright trunk position on lower extremity joint moments and powers in older adults. Contrary to our hypotheses, step length and trunk position did not affect ankle or hip joint kinetics, and did not reverse the age-related distal to proximal shift of kinetics. These findings can serve as a starting point for investigating alternative explanations for the shift, such as the role of physical activity with age, or age-related differences in muscle properties and muscle structure.

In the present study, older adults self-selected a similar step length compared to young adults walking at 1.3 m/s. Therefore, step lengths in the 1.3-SS and 1.3-SL conditions were not different, and there were no differences in joint moments or powers when older adults walked with a step length similar to young adults vs. walking with a self-selected step length. In contrast, older adults are typically observed to walk with shorter step lengths, compared to young adults (Judge et al., 1996a; Kerrigan et al., 1998; Winter

et al., 1990). However, Silder et al. (2013) reported similar step lengths between active young and older adults. It is possible that the high physical activity level of both the young and older adults in the present study influenced the self-selected step lengths. Despite the similar self-selected step lengths, older adults in this study displayed the characteristic age-related distal to proximal shift in lower-extremity joint moments and powers, consistent with the age-related differences observed in other studies (Cofré et al., 2011; DeVita & Hortobagyi, 2000b; Silder et al., 2008). The presence of these age-related kinetic differences, in the absence of spatiotemporal differences, indicate that using the same step length as young adults does not reverse or reduce the age-related distal to proximal shift in older adults. It is therefore unclear why older adults typically take shorter steps compared to young adults. Given that these very fit, highly active older adults had similar self-selected step lengths as young adults, we speculate that shorter steps may be more a function of fitness than age and could be adaptations to offset fear of falling by increasing the time spent in double-support (Maki, 1997).

Trunk position also did not affect joint kinetics in these older adults. Despite a ~3 degree increase in peak trunk extension and ~3 degree decrease in peak trunk flexion (i.e. older adults walked with a more upright posture), hip flexion and extension were similar between conditions (Fig. 4), and there were no differences in hip or ankle kinetics between conditions, indicating trunk position also did not reverse or reduce the age-related distal to proximal shift in these participants. Alternatively, pelvis position, rather than trunk position, is also suggested to affect age-related differences in hip position (Judge et al., 1996b; Kerrigan et al., 1998), and in effect, hip kinetics, and kinematically, has a more direct effect on hip angle than the trunk. Manipulating the pelvis angle during gait is

challenging, but could be investigated with optimal control simulations (Miller et al., 2015).

To summarize, the presence of age-related kinetic differences, in the absence of spatiotemporal differences, and the lack of differences between walking with and without greater trunk extension, indicate that neither step length nor trunk position explain the age-related distal to proximal shift. Therefore, alternative sources for this shift should be explored. For example, age-related differences in the length-tension and force-velocity relationships (Thom et al., 2005), and age-related differences in muscle architecture, such as gastrocnemius fascicle length and pennation angle (Narici et al., 2003), may help to explain the age-related reduction in ankle plantarflexion moment and power generation. Further, since these properties can be influenced by physical activity (Blazevich, 2006), the role of physical activity in mediating the distal to proximal shift in gait kinetics should be investigated. Boyer et al. reported that moderately-active older adults (~11,000 steps/day, no impact/jumping sports or jogging/running) still exhibited the distal to proximal shift in kinetics, but to our knowledge the effect of very high fitness levels (e.g. master's athletes) on the age-related shift in kinetics compared to less fit older adults in unknown. This gap could further clarify if the characteristic age-related shift in joint moments and powers is maladaptive, or a normal, healthy consequence of aging.

This study's scope is limited in that the participants in this study were healthy and highly physically active, which may limit the ability to directly compare these results with those collected from a more typical, less physically active, older adult population, or to older women, who were not studied. However, this highly active subject population allowed for interpretations to be made about the effect of age without the potentially

confounding effect of physical activity or chronic illness. Additional limitations are the lack of a major step length manipulation, and the lack of statistical comparisons with the young adult group. Previous studies on gait mechanics in highly fit and active older adults are limited but suggest step length differences exist between physically active young and older adults (Boyer et al., 2012; Savelberg et al., 2007). Therefore we did not anticipate that these older adults would self-select step lengths similar to young adults. As a result, conclusions made from the present study may be limited in their generalizability to older adults who self-select a step length similar to young adults. These older adults may exhibit kinetic differences when tasked with walking using an untested step length, but that step length would deviate substantially from the self-selected step length of both young and older adults in this study. We note that some subjects did indeed use different step lengths between the 1.3-SS and 1.3-SL conditions (Table 2), even though the average step lengths in both conditions were similar.

Statistical comparisons between young and older adults were not performed because the age-related distal to proximal shift of moments and powers is well-documented, even in high functioning older adults (Cofré et al., 2011; DeVita & Hortobagyi, 2000b). Additional comparisons would reduce the statistical power of the planned within-subjects comparisons and would be unlikely to produce new insights given the trends observed between young and older adults (Fig. 2 and 3).

A final limitation of note is that trunk position was not tightly controlled or systematically manipulated, and it is possible that the difference in trunk position was not large enough to elicit a kinetic response at the hip. Older adults were tasked with "walking while keeping the trunk as upright as possible". This design was intentional in order to

mimic an instruction that may be given in a typical clinical setting or as a public health recommendation. Although we did not directly control for the change in trunk position from 1.3-SS to 1.3-Trunk, kinematic results confirm that older adults did walk with a more upright trunk posture in the 1.3-Trunk condition.

3.5 Conclusion

Step length and trunk position did not reverse the age-related distal to proximal shift of joint moments and powers. Specifically, neither trunk position nor step length explain the smaller ankle kinetics and larger hip kinetics observed in older adults compared to young adults; this distal to proximal shift in joint kinetics was seen even when older adults preferred the same step length as young adults, even when they deliberately walked with the same step length as young adults, and even when they deliberately walked with an upright trunk. Alternative sources for the age-related shift in joint kinetics should be explored, such as muscle mechanical properties and muscle architecture. Future work should also address the role of physical activity in delaying or minimizing the distal to proximal redistribution of joint kinetics.

Table 3.1. Mean ± standard deviation (p) of outcome variables in the 1.3-SS, 1.3-SL, and 1.3-Trunk conditions.

	1.3-SS	1.3-SL (p = 1.3-SS vs. 1.3-SL)	1.3-Trunk (p = 1.3-SS vs. 1.3-Trunk)
Peak Joint Moments (Nm/kg)			
Hip Extensor	0.76 ±0.19	0.80 ± 0.17 (p = 0.07, Δ = 0.29)	0.75 ± 0.16 (p = 0.20, Δ = 0.07)
Hip Flexor	-0.52 ± 0.17	-0.45 ± 0.13 (p = 0.02, Δ = 0.21)	-0.42 ± 0.13 (p = 0.02, Δ = 0.29)
Ankle Plantarflexor	1.43 ± 0.16	1.45 ± 0.10 (p = 0.15, Δ = 0.11)	1.39 ± 0.31 (p = 0.20, Δ = 0.22)
Joint Moment Impulse (Nm/kg*s)			
Hip Extensor	0.07 ± 0.02	0.07 ± 0.02 (p = 0.17, Δ = 0.00)	0.07 ± 0.02 (p = 0.25, Δ = 0.00)
Hip Flexor	-0.04 ± 0.02	-0.03 ± 0.01 (p = 0.12, Δ = 0.33)	-0.04 ± 0.02 (p = 0.13, Δ = 0.00)
Ankle Plantarflexor	0.17 ± 0.03	0.19 ± 0.02 (p = 0.06, Δ = 1.00)	0.18 ± 0.02 (p = 0.04, Δ = 0.50)
Peak Joint Powers (W/kg)			
Hip Positive 1 (~15% gait cycle)	0.95 ± 0.30	1.23 ± 0.49 (p = 0.01, Δ = 0.72)	1.12 ± 0.55 (p = 0.06, Δ = 0.44)
Hip Positive 2 (~60% gait cycle)	0.83 ± 0.24	0.79 ± 0.15 (p = 0.09, Δ = 0.06)	0.87 ± 0.20 (p = 0.18, Δ = 0.06)
Ankle Positive	2.44 ± 0.54	2.40 ± 0.39 (p = 0.18, Δ = 0.08)	2.43 ± 0.46 (p = 0.07, Δ = 0.02)
Joint Work (J/kg)			
Hip Positive	0.09 ± 0.03	0.09 ± 0.03 (p = 0.12, Δ = 0.00)	0.10 ± 0.03 (p = 0.03, Δ = 0.33)
Ankle Positive	0.07 ± 0.02	0.07 ± 0.02 (p = 0.06, Δ = 0.00)	0.07 ± 0.02 (p = 0.05, Δ = 0.00)

Table 3.2. Walking speed and step lengths for each older adult participant in the 1.3-SS and 1.3-SL conditions

	Speed (m/s)		Step Length (m)	
Participant	1.3-SS	1.3-SL	1.3-SS	1.3-SL
1	1.30	1.29	0.78	0.76
2	1.33	1.32	0.75	0.77
3	1.30	1.29	0.77	0.78
4	1.31	1.29	0.69	0.70
5	1.33	1.33	0.70	0.71
6	1.28	1.28	0.72	0.72
7	1.29	1.29	0.69	0.70
8	1.29	1.29	0.69	0.68
9	1.32	1.28	0.70	0.72
10	1.29	1.29	0.72	0.73
11	1.31	1.30	0.79	0.79
12	1.29	1.31	0.65	0.74
Mean	**1.30**	**1.30**	**0.72**	**0.73**
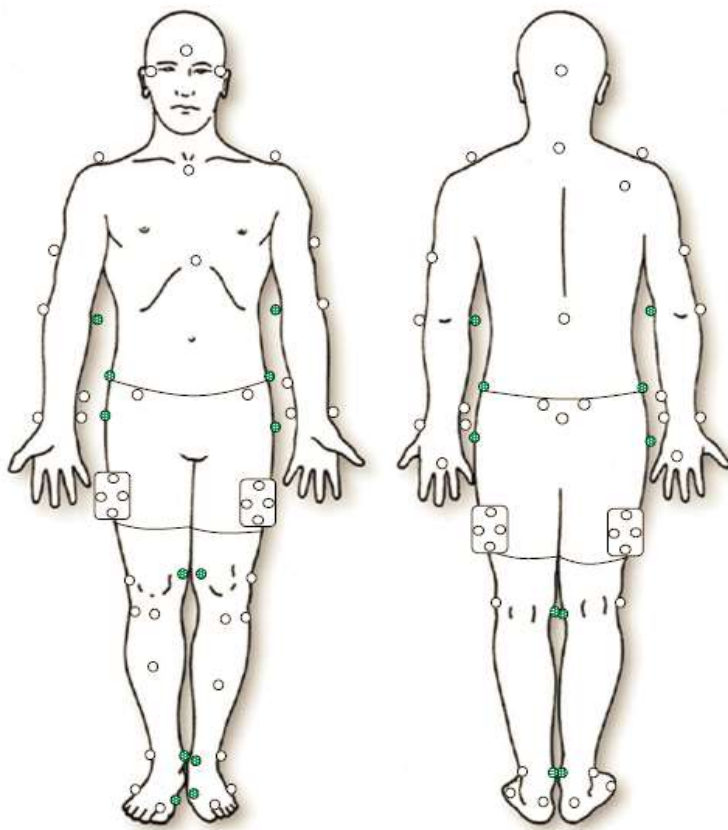
Figure 3.1. Experimental marker set-up
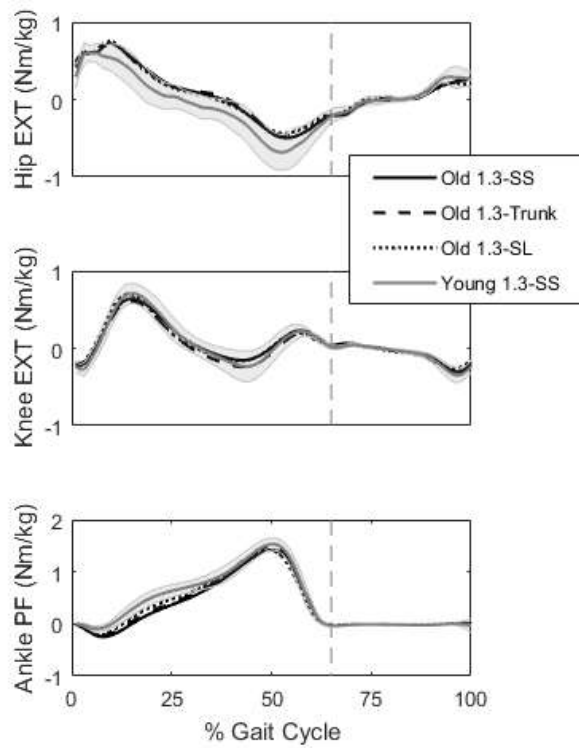
Figure 3.2. Mean sagittal plane joint moments for the hip (top), knee (middle), and ankle (bottom) in older adults during walking in the 1.3-SS- (solid black), 1.3-SL (dotted black), and 1.3-Trunk (dashed black) conditions. Positive values indicate extensor torque, negative values indicate flexor torque. The stride begins and ends at heel strike. The young adult mean and standard deviation for the 1.3-SS condition is included for reference (solid gray and shaded gray). Vertical dashed line indicates toe-off.
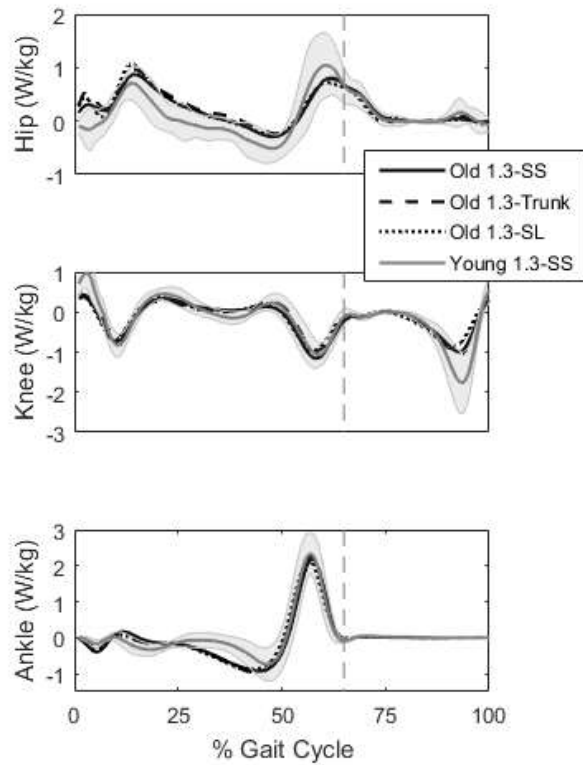
Figure 3.3. Mean sagittal plane joint powers at the hip (top), knee (middle), and ankle (bottom) in older adults during walking in the 1.3-SS (solid black), 1.3-SL (dotted black), and 1.3-Trunk (dashed black) conditions. Positive work indicates mechanical energy generation and negative work indicates mechanical energy absorption. The stride begins and ends at heel strike. The young adult mean and standard deviation for the 1.3-SS condition is included for reference (solid gray and shaded gray). Vertical dashed line indicates toe-off.
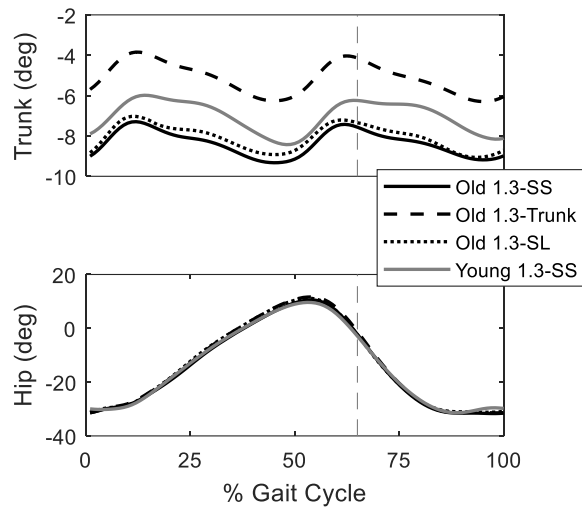
Figure 3.4. Mean sagittal plane joint angles at the trunk (top) and hip (bottom) in older adults during walking in the 1.3-SS (solid black), 1.3-SL (dotted black), and 1.3-Trunk (dashed black) conditions. Trunk angle is relative to the global coordinate system. Positive values are extension and negative values are flexion. The stride begins and ends at heel strike. The young adult 1.3-SS condition is included for reference (solid gray). Vertical dashed line indicates toe-off.

CHAPTER 4

PHYSICAL ACTIVTY DOES NOT REDUCE THE AGE-RELATED DISTAL TO PROXIMAL SHIFT OF JOINT KINETICS IN OLDER ADULTS

4.1 Introduction

Habitual physical activity contributes to the maintenance of muscle mass, structure, and function with age (Sandler et al., 1991; Visser et al., 2002). However, many older adults report declining levels of physical activity and substantial mobility challenges (Hallal et al., 2012; Milanović et al., 2013). These age-related mobility limitations may be associated with well-documented age-related differences in lower-extremity gait kinetics. Older adults exhibit smaller plantarflexor moment and power, and larger hip moment and power, compared to young adults. These age-related differences in walking kinetics are often described as a 'distal to proximal shift' of lower-extremity kinetics, and are independent of speed (Cofré et al., 2011; Kerrigan et al., 1998) and step length (Judge et al., 1996a). However, little consensus exists regarding the effect of fitness and physical activity on the distal to proximal shift. This gap in knowledge is important because determining the effect of physical activity on lower-extremity kinetics in older adults will help determine if this shift is caused by low fitness rather than age, and avoidable by any means (e.g. a high level of physical activity), or if it is a natural and unavoidable consequence of normal, healthy aging.

All studies on the distal to proximal shift have not necessarily shown the same age-related differences in specific features of hip and knee kinetics, but one of the most

consistent observations is smaller plantarflexor power generation in older adults vs. young adults, which is often attributed to plantarflexor weakness(Judge et al., 1996a; Silder et al., 2008). The relationship between muscle strength and parameters of gait kinetics is illustrated by the positive correlation between maximum isometric plantarflexor torque and peak plantarflexor power during walking (Silder et al., 2008). In fact, walking kinetics appear to be particularly sensitive to the level of available strength in the plantarflexor muscles, whereby weakness in this muscle group may contribute to decreased ankle joint moment and power generation through a smaller capacity to produce muscle forces (Kulmala et al., 2016). While it is thought that regular physical activity helps to maintain muscle strength and mobility in older adults (Sandler et al., 1991; Visser et al., 2002), physical activity levels decline with age (Hallal et al., 2012; Milanović et al., 2013). Therefore, it is possible that physical activity level, independent of age, is at least partially contributing to the age-related distal to proximal shift in gait kinetics.

The effect of physical activity on gait kinetics in older adults is inconclusive. For example, the age-related distal to proximal shift was partially attenuated in older adults who engaged in a high volume of daily walking (at least 7500 steps/day, but did no running or jumping/impact activities) compared to young adults with similar physical activity level (Boyer et al., 2012), but the age-related distal to proximal shift was still present in older adults who engaged in running activities at least twice a week for two years (Savelberg et al., 2007) and in older adults who performed at least 30 minutes of moderate activity at least twice a week for a year (Buddhadev & Martin, 2016). Further, participants in the latter two studies showed no differences in gait kinetics compared to healthy, but inactive older adults. Of note, in the aforementioned studies, the definition of 'physically active'

older adult varies substantially, which makes it difficult to infer conclusions about the effect of physical activity on gait kinetics in older adults. An alternative approach for determining the effect of physical activity on the age-related distal to proximal shift is studying master's athletes, who theoretically represent a model of successful aging (Bortz IV & Bortz, 1996; Hawkins et al., 2003). If highly physically active older adults (i.e. master's athletes) and normally physically active older adults do not exhibit differences in gait kinetics, it may indicate that physical activity does not reverse or reduce the age-related distal to proximal shift, and may further suggest that this shift is a natural and unavoidable part of aging.

Therefore, the purpose of this study was to compare ankle and hip kinetics between highly physically active and normally active older adults, which for the purposes of this study are represented by older runners and non-runners, to determine the effect of high levels of physical activity on the age-related distal to proximal shift of lower-extremity joint kinetics. We hypothesized that older non-runners would display smaller ankle kinetics and larger hip kinetics compared to older runners.

4.2 Methods

Participants

Participants were 12 male older endurance runners (67 ± 5 yrs, 1.79 ± 0.07 m, 77.3 ± 13.7 kg), and 11 male older non-runners (70 ± 3 yrs, 1.78 ± 0.06 m, 79.68 ± 10.6 kg). The minimum detectable difference with error rates of 5% for false positive and 20% for false negative was Cohen's $dz = 1.0$, which is similar to or greater than the effects in other studies on physical activity in older adults (Savelberg et al., 2007). Data for 13 male young

endurance runners are also included for reference (21 ± 3 yrs, 1.80 ± 0.04 m, 70.2 ± 7.1 kg). Young and older runners self-reported running at least 20 miles/week on a regular basis for the past year (, and training for at least one race of 10 km distance or longer in the past year. Additionally, these older runners reported on average, 34 ± 14 years of running training (range: 7-60 yrs); see Appendix A for more details. Older non-runners self-reported participating in exercise no more than twice a week for 30 minutes each session. We did not measure aerobic fitness in these participants, but members of this local older runner population including several of these same participants have been tested previously and recently, with a mean VO2max of 46 ± 6 mL/kg/min (Alfini et al., 2016). This level of aerobic fitness is classified as "excellent", in the top 90-95% of this age group, by the American College of Sports Medicine (2013). Participation in this study was limited to male participants in order to eliminate potential confounding effects of sex. Prior to enrolling in this study, individuals were screened and excluded if they reported previous injuries to the legs or back that required medical attention and/or chronic medical conditions that have been associated with gait changes such as diabetes and peripheral artery disease (Myers et al., 2016). All participants provided written informed consent to procedures approved by the University of Maryland Institutional Review Board.

Isometric Strength Testing

Isometric strength testing was performed to provide an objective measure of muscular strength and to complement self-reported measures of physical activity level. Maximal isometric strength was measured at the hip (10, 20, and 30 degrees of hip flexion), knee (0, 15, and 30 degrees of knee flexion), and ankle (-10, 0, and 10 degrees of ankle plantarflexion) using an isometric dynamometer (Biodex Medical Systems, Shirley, NY,

USA). Participants performed two sets of maximal voluntary contractions (MVC) at each angle in both the flexion and extension directions. Each trial lasted five seconds. Details of participant setup are included in Figure 1

Instrumented Gait Analysis

Participants wore shorts and their own athletic shoes. Positions of 44 retroreflective markers on the pelvis and lower limbs (Fig. 2) were captured at 200 Hz using a 13-camera motion capture system (Vicon, Oxford, UK). Force plate data were captured at 1000 Hz using 10 six degree of freedom piezoelectric force plates (Kistler, Switzerland) arranged in a single row located within a raised platform. All subjects completed five walking trials at a standard speed of 1.3 m/s with a self-selected step length. Speed was monitored using an infrared timing gate system and trials in which the average over the force plates was not within ± 3% of 1.3 m/s were excluded.

Data Processing

Five clean footstrikes isolated on a single force plate were selected for analysis for each participant. Marker position and ground reaction force data were exported to Visual3D (C-Motion, Germantown, MD, USA) and smoothed using a 4th-order dual-pass Butterworth filter with cutoff frequencies of 6 Hz and 45 Hz, respectively. A linked-segment model was created for each participant from marker positions during a standing calibration trial. Joint angles were calculated using 6DOF pose estimation and a Cardan Xyz rotation sequence (Wu & Cavanagh, 1995). Iterative Newton-Euler inverse dynamics within Visual3D were used to calculate joint powers and moments (Selbie et al., 2014). Joint moment impulse was calculated as the time integral of the moment curve, and joint

work was calculated as the time integral of the power curve. Joint moment impulse and joint work were calculated for the stance phase, defined as heel strike to toe-off (~0 – 60% of the gait cycle).

From the isometric joint strength tests, raw torque-time data were exported from the dynamometer. Peak torque at each joint was determined as the larger of the two MVC measurements during which there was no acceleration registered on the dynamometer arm. Torque data were scaled by participant mass.

Statistical Analysis

Outcome variables from the gait data were peak sagittal plane hip extensor and flexor moments and impulse, peak ankle plantarflexor moment and impulse, and peak hip and ankle positive powers and joint work. Variables were determined from each trial and averaged over trials to produce a representative value for each participant. All variables were scaled by participant mass. To test for the effect of physical activity, an independent samples one-sided *t*-test was conducted to determine differences in outcome variables between the older runners and non-runners. Bonferroni correction was used to adjust for multiple comparisons resulting in critical $p < 0.005$. To complement the significance testing, effect sizes (Δ) were calculated, based on Glass' Delta, as the difference in means between groups divided by the standard deviation of young adults. The standard deviation of young adults were used in this calculation in order to interpret effect sizes in reference to between-subjects variance that is presumably normal and healthy.

4.3 Results

Isometric Strength

Older runners displayed 23, 19, and 19% greater hip extensor torque at 10, 20, and 30 degrees of hip extension, respectively, and similar hip flexor torque at all hip angles, compared to older non-runners (Fig. 3A). At the knee, older runners displayed 39 and 38% greater knee extensor torque at 15 and 30 degrees of knee flexion compared to older non-runners, respectively. Older runners and non-runners displayed similar knee extensor torque at 0 degrees of knee flexion, and similar flexor torque at all knee angles (Fig. 3B). At the ankle, older runners displayed similar plantarflexor torque at-10, 0, and 10 degrees of plantarflexion, and displayed 37, 44, and 41% 40% greater dorsiflexor torque at -10, 0, and 10 degrees of plantarflexion, respectively, compared to older non-runners (Fig. 3C). In summary, the older runners in this study were generally "stronger" (on the basis of maximum voluntary isometric torque) than the older non-runners in hip extension, knee extension, and ankle dorsiflexion, but not in ankle plantarflexion.

Gait Mechanics

Average speed and step lengths for each group were: older runners (1.30 ± 0.02 m/s and 0.72 ± 0.04 m), older non-runners (1.30 ± 0.01 m/s and 0.72 ± 0.04 m), and young runners (1.31 ± 0.02 m/s and 0.73 ± 0.03 m).

Although there were some visually-evident differences between older runners and non-runners in features of the joint moments and powers (Fig. 4 and 5), there were no significant differences in the direction of the tested hypotheses between older runners and non-runners for any outcome variables (all $p > 0.005$, Table 1).

Differences in joint kinetics between young and older adults were consistent with the distal to proximal shift in joint moments and powers reported elsewhere (e.g. DeVita & Hortobagyi, 2000). Older runners and non-runners each displayed 19% greater peak hip power in early stance (Δ = -0.46), 46 and 19% smaller peak hip power in late stance (Δ = 0.76 and 0.35), respectively, and older runners displayed 36% greater positive hip work (Δ = -0.98), compared to young runners (Fig. 4 and 5). Older runners and non-runners displayed 8 and 7% smaller peak ankle moments (Δ = 0.82 and 0.77, respectively), and 5% smaller peak ankle power (Δ = 0.27), compared to young runners. Both groups exhibited 17% smaller ankle plantarflexor moment impulse (Δ = 1.20), and older runners generated 13% less positive work at the ankle (Δ = 0.67), compared to young runners (Fig. 4 and 5).

4.4 Discussion

The purpose of this study was to determine the effect of running-related physical fitness on joint moments and powers during walking in older adults. Older runners demonstrated higher levels of fitness compared to older non-runners. These older runners self-reported engaging in a high volume of aerobic exercise (run $\geq$ 20 miles/week) for at least the past seven years, with 11 participants (92% of the older runners) reporting 30+ years of such training, and several of these same participants previously exhibiting VO2max scores classified as "excellent" (Alfini et al., 2016) by the American College of sports Medicine (2013). Additionally, the older runners displayed generally greater isometric leg strength compared to older non-runners. We believe this is the "fittest"

cohort, on the basis of running-related fitness, that the distal-to-proximal shift in lower extremity joint kinetics has been investigated in to date. However, contrary to our hypothesis, older runners did not exhibit smaller hip kinetics or larger ankle joint kinetics compared to older non-runners. We found moderate to large effect sizes for hip flexor moment (Δ = -0.84), hip flexor impulse (Δ = -0.90), hip extensor impulse (Δ = 1.14), and positive hip work (Δ = 0.84), but these differences were not in the direction of the *a priori* hypotheses. The present findings suggest that the characteristic age-related distal to proximal shift persists despite high levels of running related fitness and therefore may be an unavoidable part of natural aging.

Older runners displayed greater isometric hip extensor, knee extensor, and dorsiflexor strength compared to older non-runners. Although leg strength has been shown to partially mediate age-related differences in gait kinetics (Hortobágyi et al., 2016), older runners in the present study did not display differences in gait kinetics compared to older non-runners. These findings are consistent with previous literature reporting no differences in gait kinetics between moderately active and sedentary older adults (Buddhadev & Martin, 2016). Notably, in the present study, older runners and non-runners displayed similar peak isometric plantarflexor strength, suggesting that although running may promote greater hip and knee extensor strength, it is not sufficient for promoting greater ankle extensor strength (i.e. plantarflexion strength) in older adults. Maximum isometric plantarflexor torque has been previously associated with peak positive plantarflexor power during gait (Silder et al., 2008). Therefore, the similarities between older runners and non-runners in ankle joint kinetics during walking may be related to the observed similarities in isometric ankle plantarflexor strength between groups. To our knowledge, no studies

have determined the effect of plantarflexor specific strength training on gait kinetics in healthy older adults. A history of running here did not elicit greater ankle kinetics during walking, but it is unknown whether an alternative mode of exercise (e.g. strength training) would be effective in reversing or reducing the age-related distal to proximal shift.

The persistence of the age-related shift in kinetics, despite previous attempts at manipulating walking speed (Cofré et al., 2011), step length (Chapter 3), trunk kinematics (Chapter 3), and physical activity level, may indicate that these differences are secondary to adaptations at the muscle level. For example, age-related differences in muscle properties, such as maximum isometric force, and age-related differences in muscle architecture, such as gastrocnemius fascicle length and pennation angle (Narici et al., 2003), may help to explain the age-related reduction in ankle plantarflexion moment and power generation. Since sagittal-plane joint moments at the hip and ankle are due primarily to muscle forces during walking, future studies should investigate the distal to proximal shift in kinetics at the muscle level.

This study's scope is limited in that our definition of 'highly physically active' pertains only to endurance runners. It is possible that individuals who participate in other modes of physical activity, such as strength or power training would exhibit different ankle and hip joint kinetics. We chose to recruit runners because many active older adults choose running or walking as their primary mode of physical activity, both of which are easily accessible activities that may be suggested by clinicians for maintaining physical activity with age, and also simply for practicality (the local communities have a large number of older runners). Additionally, older non-runners in this study were healthy, with no history of chronic medical conditions, and were only excluded based on their physical activity

history if they participated in exercise more than 30 minutes twice a week during the past year. The average older adult population may engage in less physical activity than the inclusion criteria for the non-runners. However, these criteria were selected to avoid the potentially confounding effects of chronic disease and sedentary living, which are associated with kinetic changes similar to the characteristic age-related shift in joint moments and powers (Myers et al., 2016). It is also possible that the standard speed of 1.3 m/s was not fast enough to elicit kinetic differences between older adult groups, as several studies have shown that age-related differences in hip and ankle kinetics are exacerbated at faster walking speeds (Cofré et al., 2011; Kerrigan et al., 1998). However, this shift has been observed in speeds as slow as 1.0 m/s (Cofré et al., 2011), and the standard speed of 1.3 m/s was selected to be representative of a walking speed that is easily achievable for many older adults, regardless of their physical activity level. Finally, statistical comparisons between young and older adults were not performed here, as our interested was primarily in the effect of fitness / physical activity in older adults. Additional comparisons would reduce the statistical power of planned comparisons, and the age-related distal to proximal shift of joint moments and powers is well-documented in high-functioning older adults (Cofré et al., 2011; DeVita & Hortobagyi, 2000b). Therefore, it did not seem necessary to conduct similar comparisons here as they would be unlikely to produce new insights.

4.5 Conclusion

Physical activity in the form of running at least 20 mi/wk and training for at least one race per year did not reduce the age-related distal to proximal shift of joint moments

and powers in older adults. Therefore, it is unlikely that even high levels of running can reverse or mitigate the age-related distal to proximal shift in joint moments and powers. Rather, these results suggest that age-related differences in lower-extremity joint kinetics are an unavoidable part of natural aging even in the absence of any overt mobility limitations and the presence of a high level of running-related fitness. Future work should characterize the effects of varying modes of physical activity (i.e. strength training vs. endurance training) on kinetics in older adults. Alternative sources for the age-related shift in joint kinetics should be explored, such as age-related differences in muscle properties and muscle forces.

Table 4.1. Mean ± standard deviation of outcome variables. Positive effect size indicates the mean of the Older Runners group was greater than the mean of the Older Non-Runners group.

	Older Runners (n = 12)	Older Non-Runners (n = 11)	Statistics
Peak Joint Moments (Nm/kg)			
Hip Flexor	-0.52 ± 0.17	-0.79 ± 0.25	(p = 0.997, Δ = -0.84)
Hip Extensor	0.76 ±0.19	0.73 ± 0.11	(p = 0.629, Δ = 0.15)
Ankle Plantarflexor	1.43 ± 0.16	1.44 ± 0.19	(p = 0.538, Δ = -0.05)
Joint Moment Impulse (Nm/kg*s)			
Hip Flexor	-0.04 ± 0.02	-0.07 ± 0.03	(p = 0.997, Δ = -0.90)
Hip Extensor	0.07 ± 0.02	0.05 ± 0.02	(p = 0.992, Δ = 1.14)
Ankle Plantarflexor	0.17 ± 0.03	0.17 ± 0.02	(p = 0.538, Δ = 0.00)
Peak Joint Powers (W/kg)			
Hip Positive 1 (~15% gait cycle)	0.95 ± 0.30	0.95 ± 0.38	(p = 0.501, Δ = 0.00)
Hip Positive 2 (~60% gait cycle)	0.83 ± 0.24	1.09 ± 0.35	(p = 0.975, Δ = -0.40)
Ankle Positive	2.44 ± 0.54	2.45 ± 0.35	(p = 0.506, Δ = -0.01)
Joint Work (J/kg)			
Hip Positive	0.09 ± 0.03	0.06 ± 0.02	(p = 0.986, Δ = 0.85)
Ankle Positive	0.07 ± 0.02	0.08 ± 0.01	(p = 0.849, Δ = -0.41)

Figure 4.1. Participant setup for isometric strength testing. A) Hip: participants were supine with a Velcro immobilizer strap placed high across the pelvis. The dynamometer attachment was placed on the distal portion of the thigh with the lever arm parallel to the thigh and the axis of rotation in line with the greater trochanter. B) Knee: participants were seated in the chair with the back angle set at 85 degrees and a Velcro immobilizer strap placed across the distal thigh. Immobilizer seatbelts were used across the chest and hips. The dynamometer attachment was placed on the distal aspect of the shank with the lever arm parallel to the shank and the axis of rotation in line with the axis of rotation of the knee joint. C) Ankle: participants were seated in the chair with the back angle set at 85 degrees and the knee flexed at 45 degrees and supported with a separate attachment under the distal end of the thigh. The participant's foot was securely strapped to the footplate and positioned so the axis of rotation was in line with the lateral malleolus.
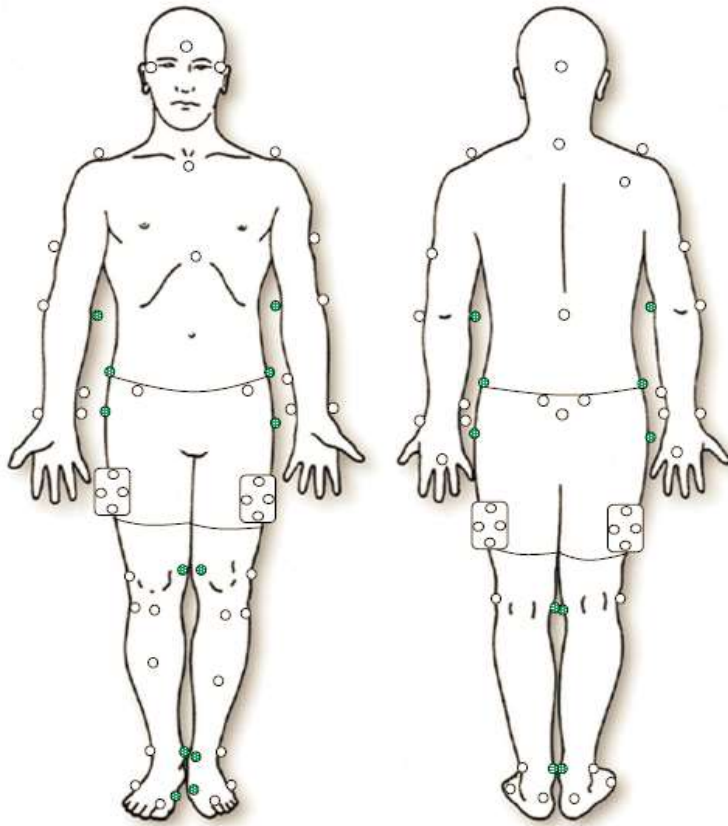
Figure 4.2. Experimental marker set-up
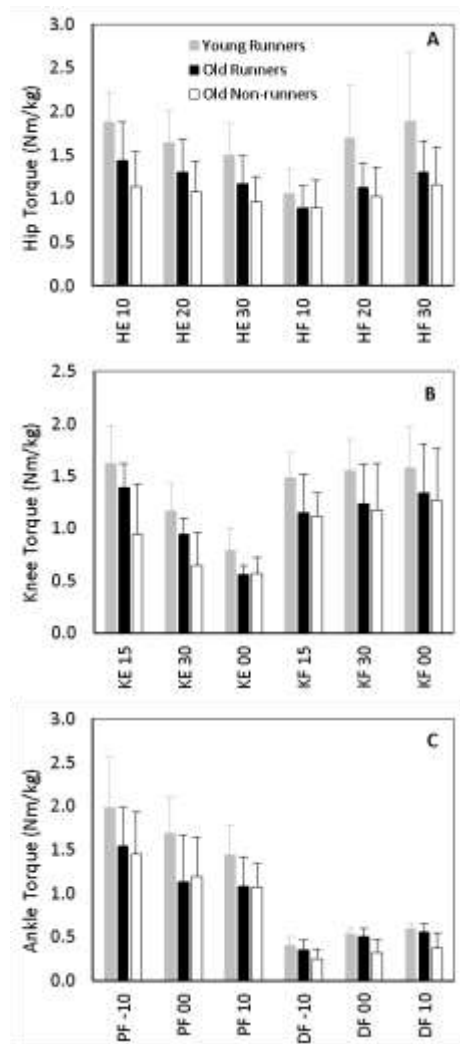
Figure 4.3. Mean and standard deviation of maximum voluntary torque at three joint angles for A) hip extension (HE) and hip flexion (HF), B) knee extension (KE) and knee flexion (KF), and C) ankle plantarflexion (PF) and ankle dorsiflexion (DF). Mean and standard deviation for young runners are included for reference (gray bars).
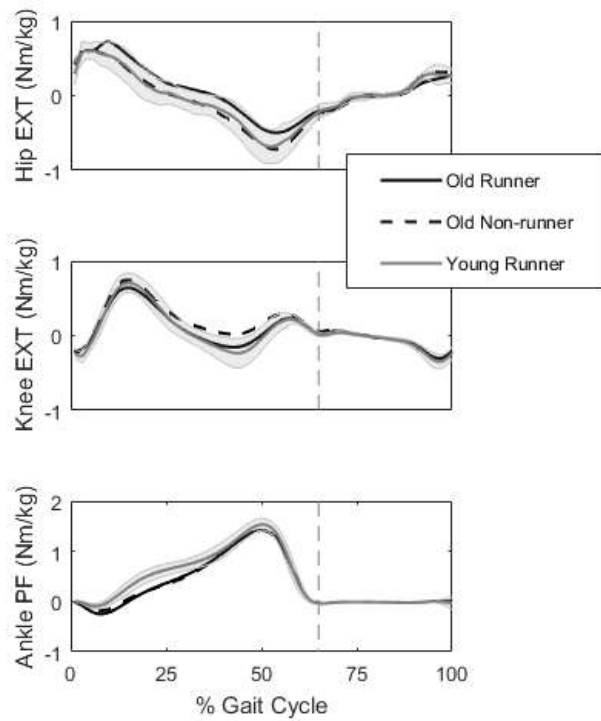
Figure 4.4. Mean sagittal plane joint moments for the hip (top), knee (middle), and ankle (bottom) in older runners (solid black) and older non-runners (dashed black) during walking at 1.3 m/s. Positive values indicate extensor torque, negative values indicate flexor torque. The stride begins and ends at heel strike. Mean (solid gray) and standard deviation (shaded gray) for the young runners is included for reference. Vertical line indicates toe-off.
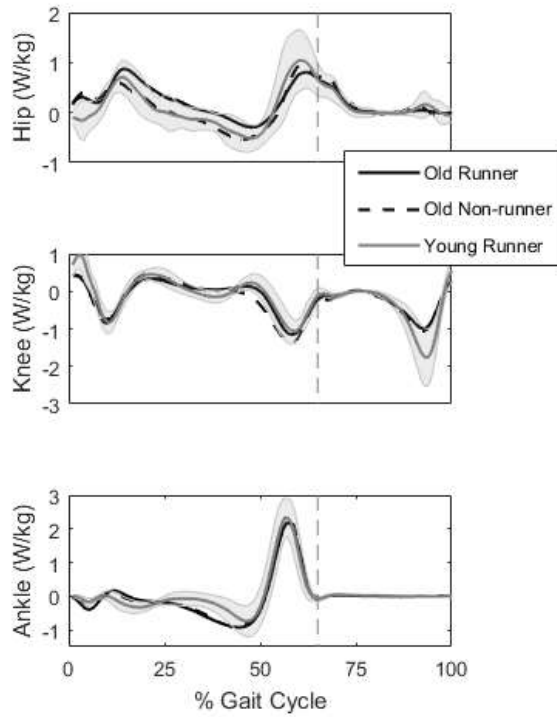
Figure 4.5. Mean sagittal plane joint powers at the hip (top), knee (middle), and ankle (bottom) in older runners (solid black) and older non-runners (dashed black) during walking at 1.3 m/s. Positive work indicates mechanical energy generation and negative work indicates mechanical energy absorption. Mean (solid gray) and standard deviation (shaded gray) for the young runners is included for reference. Vertical line indicates toe-off.

CHAPTER 5

THE AGE-RELATED DISTAL TO PROXIMAL SHIFT OF KINETICS IN LOWER-EXTREMITY MUSCLE FORCES

5.1 Introduction

Older adults display differences in lower-extremity gait kinetics compared to young adults. These differences are characterized by smaller ankle plantarflexor and hip flexor moments and powers, and greater hip extensor moment and power, and are often referred to as a 'distal to proximal shift' of kinetics (DeVita & Hortobagyi, 2000a). Understanding the mechanisms behind these age-related differences would help to determine if this shift is a maladaptation that should be corrected, or if this is an unavoidable part of healthy, natural aging. However, the sources of this shift are, unknown. Several sources have been proposed, such as age-related differences in walking speed (Cofré et al., 2011), step length (Chapter 3), trunk kinematics (Chapter 3), and physical activity level (Chapter 4). However, when these factors are systematically controlled for, characteristic age-related differences in joint kinetics persist.

The persistence of the age-related shift in gait kinetics, despite previous attempts at manipulating spatiotemporal and kinematic variables, may indicate that age-related adaptations are occurring at the muscle level (e.g. muscle architecture and force generating ability). Features of muscle architecture such as fascicle length and pennation angle influence the amount of force a muscle is able to produce. Greater pennation angles generate more force compared to muscles with relatively smaller pennation angles by 1) allowing more fascicles to act in parallel along the tendon, i.e. a larger physiological cross

sectional area (PCSA), 2) operating closer to the length at which maximum force can be produced by shortening less for a given tendon excursion, and 3) operating closer to the velocity at which maximum force can be produced by reducing fascicle velocity for a given whole muscle shortening velocity (Blazevich, 2006). Similarly, muscles with relatively longer fascicles produce greater force due to the simultaneous contraction of serially arranged sarcomeres by (Spector et al., 1980). For a given whole-fascicle shortening velocity, longer fascicles (with relatively more serially arranged sarcomeres) can generate greater force by operating closer to the velocity at which maximum force can be produced. The gastrocnemius muscle is a common focus in studies investigating the effects of age on muscle architecture. Effects of age on gastrocnemius muscle architecture are inconsistent, but suggest older adults often have shorter gastrocnemius muscle fascicle lengths (Narici et al., 2003), and smaller pennation angles (Morse et al., 2005; Narici et al., 2003) compared to young adults. Therefore, age-related differences in muscle fascicle length and pennation angle may contribute to the distal to proximal shift of joint moments and powers by limiting the ability of muscles to generate force.

It is unclear if age-related differences in architecture are due to aging, per se, or if they are a result of reduced physical activity with age. However, older adult endurance runners represent a successful model of aging due to their long-term participation in exercise (Bortz IV & Bortz, 1996; Hawkins et al., 2003). Investigating the sources of age-related differences in kinetics in this population can account for the potential confounding relationship of physical activity and allow for interpretation of differences based on the effect of age.

Previous studies have estimated muscle forces in older adults to determine muscle contributions to support and forward progression of the center of mass (Lim et al., 2013; Schloemer et al., 2016), but the age-related distal to proximal shift has not been specifically assessed at the muscle level (e.g. through analysis of peak muscle forces). Examining this phenomenon at a new level of complexity could inform the sources of the age-related shift in joint kinetics, and whether this shift is due to negative and/or preventable changes in muscle properties. Further, this gap in knowledge is important because identification of the specific muscles contributing to age-related differences in kinetics may be clinically meaningful for the development of training programs to increase or preserve mobility in older adults. Therefore, the objectives of this study were to compare 1) lower-extremity muscle forces, and 2) medial gastrocnemius fascicle length and pennation angle between young and older adults with a relatively high level of fitness and physical activity (runners).

5.2 Methods

Participants

Participants were 10 older adult men (67 ± 5 yrs, 1.78 ± 0.07 m, 75.8 ± 14.6 kg) and 10 young adult men (20 ± 2 yrs, 1.80 ± 0.05 m, 70.9 ± 8.3 kg). The minimum detectable difference with error rates of 5% for false positive and 20% for false negative was $dz = 1.2$, which is similar to or greater than the effects in other studies on muscle architecture in older adults (Narici et al., 2003) and muscle forces (Schloemer et al., 2016). Both young and older adults were highly physically active, and self-reported running at least 20 miles/week and training for at least one race of 10 km distance or longer in the past year. The older adults each self-reported that they had been running for at least the past seven

years, with most reporting 20-30 years or more of training. Participation in this study was limited to male participants in order to eliminate potential confounding effects of sex. Prior to enrolling in this study, individuals were screened and excluded if they reported previous injuries to the legs or back that required medical attention and/or chronic medical conditions that have been associated with gait changes such as diabetes and peripheral artery disease (Myers et al., 2016). All participants provided written informed consent to procedures approved by the University of Maryland Institutional Review Board.

Instrumented Gait Analysis

Participants wore shorts and their own athletic shoes. Positions of 44 retroreflective markers on the pelvis and lower limbs (Fig. 1) were captured at 200 Hz using a 13-camera motion capture system (Vicon, Oxford, UK). Force plate data were captured at 1000 Hz using 10 six degree of freedom piezoelectric force plates (Kistler, Switzerland) arranged in a single row located within a raised platform. Participants completed five walking trials at a standard speed of 1.3 m/s with a self-selected step length. Speed was monitored using an infrared timing gate system and trials in which the average over the force plates was not within ± 3% of 1.3 m/s were excluded. Surface electromyography (EMG) was collected for seven sites: tibialis anterior (TA), soleus (SOL), medial gastrocnemius (GA), vastus lateralis (VL), rectus femoris (RF), biceps femoris (BF), and semitendinosus (SMT). EMG data were collected at 2000 Hz using Trigno wireless EMG system (Delsys Inc., Natick, MA, USA). Electrode placement sites were identified using SENIAM guidelines (Hermens et al., 1999), and were prepared for data collection using a disposable razor and alcohol swab to prior to attaching the adhesive sensor.

Ultrasound Measurement of Muscle Architecture

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Participants were seated with their ankle at 90 degrees and knee at 0 degrees (Karamanidis & Arampatzis, 2005). The medial head of the gastrocnemius was imaged while the muscle was relaxed, using a Lumify (Philips, USA) L12-4 broadband linear array transducer (12-4 MHz, 34 mm scanning length, M-mode). The ultrasound probe was oriented along the mid-sagittal axis of each muscle and secured in place using flexible support and adhesive tape to ensure the same region of the muscle was capture in each image. Three images were collected for each participant.

Data Processing

Five clean footstrikes isolated on a single force plate were selected for analysis for each participant. Marker position and ground reaction force data were exported to Visual3D (C-Motion, Germantown, MD, USA) and smoothed using a 4th-order dual-pass Butterworth filter with cutoff frequencies of 6 Hz and 45 Hz, respectively. A linked-segment model was created for each participant from marker positions during a standing calibration trial. Joint angles were calculated using 6DOF pose estimation and a Cardan Xyz rotation sequence (Wu & Cavanagh, 1995). Iterative Newton-Euler inverse dynamics within Visual3D were used to calculate joint moments (Selbie et al., 2014).

Estimation of Muscle Forces

For each participant, joint angles and joint moments from the inverse dynamics analysis were used to estimate lower-extremity muscle forces of 43 muscles (Table 1). Muscle forces were estimated using the sequential quadratic programming algorithm in the Matlab Optimization toolbox (The Mathworks, MA, USA). For each step of the gait cycle, the muscle forces that (i) minimized the sum of squared muscle stresses subject to the

bounds imposed by each muscle's force-length-velocity properties, and (ii) matched the inverse dynamics moments at the frontal and sagittal plane of the hip, sagittal plane of the knee, and sagittal plane of the ankle, were calculated. A modification of the Hill equation (Pedotti et al., 1978) was used to calculate the maximum force that could be generated at each time step based on the current length and velocity of each muscle. Optimal fascicle length and PCSA for each muscle was obtained from Ward et al. (2009). Activation dynamics was neglected but has a negligible effect when predicting muscle forces during walking (Anderson & Pandy, 2001) and was not necessary here since muscle excitations were not an outcome. Muscle moment arms and muscle lengths were derived from Menegaldo et al. (2004), and were calculated as a function of joint angle at each time step. EMG was used for temporal validation of estimated muscle forces.

Muscle Architecture

ImageJ software (National Institutes of Health, Bethesda, MD) was used to measure fascicle length and pennation angle from the ultrasound images. Fascicle length was defined was the length of the fascicle between its insertions on the superficial and deep aponeuroses (Fig. 1). If the fascicle extended off the acquired ultrasound image, the length was estimated by extrapolating the fascicular path and aponeurosis. Errors from this approximation method are small, particularly when there is negligible curvature of the muscle belly, such as when the muscle is at rest (Narici et al., 2003). Pennation angle was measured as the angle of insertion of muscle fascicles into the deep aponeurosis (Fig. 1)

Statistical Analysis

Outcome variables from the musculoskeletal model were peak muscle forces for 12 muscles or muscle groups: iliacus, psoas, gluteus maximus, gluteus medius, vasti (lateralis, medius, and intermedius), medial hamstrings (semitendinosus, semimembranosus), rectus femoris, biceps femoris (long and short head), tibialis anterior, soleus, lateral gastrocnemius, and medial gastrocnmemius. Outcome variables for muscle architecture were medial gastrocnemius fascicle length and pennation angle. Independent two-tailed *t*-tests were used to determine differences between young and older adults. Critical was set at $p < 0.05$. Due to the exploratory nature of this study, *p* values were not adjusted for multiple comparisons. Effect sizes (Δ) were calculated, based on Glass' Delta, as the difference in means between conditions divided by the standard deviation of young adults. The standard deviation of young adults were used in this calculation in order to interpret effect sizes in reference to between-subjects variance that is presumably normal and healthy.

5.3 Results

Gait Mechanics in Young and Older Adults

Average speed and step lengths for the older adult group were 1.30 ± 0.02 m/s and 0.72 ± 0.04 m, respectively, and average speed and step length for young adult group were 1.31 ± 0.02 m/s and 0.73 ± 0.03 m, respectively.

Differences in joint kinetics between young and older adults were consistent with the distal to proximal shift in joint moments and powers reported elsewhere (e.g. DeVita & Hortobagyi, 2000): older adults displayed 46 and 55% smaller peak hip flexor moment

(Δ = -0.91) and impulse (Δ = -0.97), respectively, 33% greater hip extensor moment impulse during stance (Δ = -1.00), and 40% more positive work at the hip during stance (Δ = -0.98), compared to young adults (Fig. 2). Additionally, older adults exhibited an 8 and 16% smaller peak ankle plantarflexor moment (Δ = 0.82) and impulse (Δ = 1.19) and generated 13% less positive work at the ankle (Δ = 0.67) compared to young adults (Fig. 2).

Age-related differences in joint kinematics were also similar to previous reports (DeVita & Hortobagyi, 2000a; Judge et al., 1996b; Silder et al., 2008). Older adults displayed similar peak hip extension and hip flexion during stance compared to young adults (Δ = 0.56 and Δ = 0.69, respectively). At the knee, older adults displayed 61% smaller peak knee extension during stance compared to young adults (Δ = 1.05), and 11% more flexion in terminal stance than young adults (Δ = 1.50). At the ankle, older adults displayed 25% more dorsiflexion during stance (Δ = 1.10), and 50% less plantarflexion in terminal stance (Δ = 1.17) compared to young adults (Fig. 3).

Muscle Forces

Peak muscle forces for gluteus maximus (1.14 ± 0.30 BW vs. 0.88 ± 0.15 BW, $p = 0.02$, Δ = -1.76) and gluteus medius (2.13 ± 0.49 BW vs. 1.68 ± 0.21 BW, $p = 0.02$, Δ = -2.16) were greater in older adults compared to young adults (Fig. 3). Peak muscle forces for iliacus (0.53 ± 0.19 BW vs. 0.77 ± 0.21 BW, $p = 0.02$, Δ = 1.12), psoas (0.33 ± 0.13 BW vs. 0.49 ± 0.13 BW, $p = 0.01$, Δ = 1.28), medial hamstrings (0.86 ± 0.14 BW vs. 1.22 ± 0.47 BW, $p = 0.03$, Δ = 0.78), medial gastrocnemius (1.30 ± 0.19 BW vs. 1.57 ± 0.31 BW, $p = 0.04$, Δ = 0.84), and lateral gastrocnemius (0.37 ± 0.13 BW vs. 0.57 ± 0.23 BW, $p = 0.03$, Δ = 0.85) were smaller for older adults compared to young adults (Fig. 2). No

difference in peak muscle force between older and young adults was found for biceps femoris (0.32 ± 0.12 BW vs. 0.59 ± 0.48 BW, $p = 0.10$, $\Delta = 0.56$), rectus femoris (0.42 ± 0.11 BW vs. 0.45 ± 0.07 BW, $p = 0.57$, $\Delta = 0.32$), tibialis anterior (0.47 ± 0.13 BW vs. 0.55 ± 0.30 BW, $p = 0.48$, $\Delta = 0.25$), or soleus (3.20 ± 0.24 BW vs. 3.20 ± 0.50 BW, $p = 0.99$, $\Delta = 0.00$). Overall, the timing of muscle forces was similar between young and older adults, and consistent with measured EMG activity (example in Fig. 4).

Muscle Architecture

Medial gastrocnemius fascicles were shorter in older adults vs. young adults (46.2 ± 4.0 mm vs. 50.3 ± 4.2 mm, $p = 0.04$, $\Delta = 0.95$; Fig. 5). Pennation angle was not different between older and young adults (22.4 ± 3.0 mm vs. 22.1 ± 1.8 mm, $p = 0.72$, $\Delta = -0.22$; Fig. 5).

5.4 Discussion

The objectives of this study were to compare 1) lower-extremity muscle forces, and 2) medial gastrocnemius fascicle length and pennation angle between young and older adults with similar (and relatively high) levels of physical activity via running, to explore the age-related distal to proximal shift in kinetics at the muscle level. The distal to proximal shift in joint kinetics was reflected in lower-extremity muscle forces as older adults produced larger forces in the gluteus medius and gluteus maximus, and smaller forces in the iliacus, psoas, and medial and lateral gastrocnemius muscles, compared to young adults. Young and older adults walked at similar speeds and with similar self-selected step lengths, indicating that age-related differences in muscle forces are independent of these

spatiotemporal parameters, and providing further support for the robustness of age-related differences in joint moments and powers. Previous studies have attempted to determine the source of age-related differences in joint moments and powers through manipulation of walking speed (Cofré et al., 2011), step length (Chapter 3), trunk kinematics (Chapter 3), and physical activity (Chapter 4) with little success. Despite previous speculations that age-related differences in joint kinetics are due to spatiotemporal factors (Judge et al., 1996a), joint kinematics (Judge et al., 1996b; Kerrigan et al., 1998) or physical activity (Boyer et al., 2012), the present findings suggest that age-related differences in gastrocnemius fascicle length and force production contribute to age-related differences in kinetics.

Gastrocnemius fascicle length was found to be shorter in older adults compared to young adults, which is consistent with previous studies in recreationally active young and older adults (Morse et al., 2005; Narici et al., 2003). The presence of shorter fascicle lengths in older adults has implications for the width of the force-length curve and related force-length-velocity properties. Shorter fascicles operate further from the velocity at which maximum force can be produced, thereby producing smaller force compared a longer fascicle for a given whole-fascicle shortening velocity (Spector et al., 1980).

It is unclear if fascicle length can be increased by any means. The effect of resistance training on fascicle length is inconclusive, although the majority of evidence suggests there is little to no effect (Blazevich, 2006). Similarly, while studies on the effects of endurance training are scarce, gastrocnemius fascicle lengths in endurance runners vs. non-runners were similar (Buchholtz, 2013). Interestingly, fascicle length has been shown to decrease with detraining, or atrophy due to injury (Narici & Cerretelli, 1998). Therefore,

while habitual activity (i.e. endurance running) in older adults may mitigate decreases in fascicle length due to age-related declines in physical activity, it is unlikely to completely reverse fascicle length changes with aging, suggesting that shorter fascicle lengths, and their effect on force production, are an unavoidable part of aging.

Gastrocnemius pennation angle was not different between young and older adults, possibly due to the present participants' history of endurance running, which may serve to preserve physiological cross sectional area (i.e. the number of fascicles arranged in parallel along the tendon) and thus pennation angle (Blazevich, 2006).. Alternatively, pennation angle could also be preserved by loss of contractile tissue coupled with an age-related increase in adipose and connective tissue (i.e. noncontractile tissue). Unlike fascicle length, pennation angle has been shown to increase with training (Morse et al., 2007). Therefore, the present similarities in pennation angle between young and older runners may be due to similar training habits, thus suggesting habitual physical activity (e.g. endurance running) may mitigate age-related differences in pennation angle. Since pennation angle is roughly proportional to isometric force production (Haxton, 1944; Lieber & Friden, 2000), preserving pennation angle in older adults may help preserve maximal force capabilities.

The present results of shorter gastrocnemius fascicle lengths and smaller gastrocnemius muscle forces suggest that this muscle may be a primary contributor to the smaller moment and power generation at the ankle. Relatedly, peak soleus force was not different between young and older adults, which further supports the gastrocnemius as the primary site of age-related adaptations contributing to differences in ankle kinetics between young and older adults. The smaller gastrocnemius force in older adults may be partially

due to its biarticularity. Although the soleus and gastrocnemius have similar fascicle lengths and moment arms, knee angle influences the length of the gastrocnemius independent of the soleus, and thus influences gastrocnemius force production (Herzog et al., 1991). For example, relatively larger knee angles result in a shorter fascicle length and less gastrocnemius force. In the present study, older adults displayed greater knee flexion which may have resulted in smaller gastrocnemius muscle forces compared to young adults. However, older adults also displayed less plantarflexion compared to young adults. This combination of knee and ankle kinematics may be a strategy to maintain fascicle length and, in effect, gastrocnemius force production. However, caution is suggested with this interpretation as it may be limited by the static optimization process which preferentially assigns forces to muscles with a favorable combination of large PCSA and long moment arm. Therefore, since soleus has a large PCSA, the age-related differences observed in medial and lateral gastrocnemius, but not soleus, may be partially due to the static optimization routine preferentially assigning force to the soleus muscle, resulting in similar soleus force and smaller gastrocnemius forces to account for the smaller plantarflexor moment in older vs. young adults. Despite this potential shortcoming, the present results warrant further investigations on the influence of age-related differences in knee kinematics on gastrocnemius force production, and thus the characteristic age-related reduction in plantarflexor moment.

In addition to age-related differences in gastrocnemius force, older adults also displayed characteristic age-related differences in hip muscle forces. Compared to young adults, older adults produced larger forces in the gluteus medius throughout stance, and in the gluteus maximus in early stance, which indicates a heavy reliance on these muscles to

generate the larger hip extensor moment observed in older vs. young adults (Cofré et al., 2011; DeVita & Hortobagyi, 2000a). Age-related differences in hip flexion during walking may be associated with differences in hip kinetics. Older adults walk with more hip flexion throughout the stance phase (DeVita & Hortobagyi, 2000a), and hip the extensors can typically produce more torque at larger hip flexion angles (Anderson et al., 2007). Additionally, older adults produced smaller peak muscle forces for the iliacus and psoas muscles compared to young adults, which are consistent with the smaller flexor moments and positive power observed in older adults during late stance observed here and elsewhere (DeVita & Hortobagyi, 2000a; Monaco et al., 2009). Previous studies have speculated that the larger hip flexion angle, greater extensor moment (and thus, smaller flexion moment) is a result of greater forward trunk lean in older adults (DeVita & Hortobagyi, 2000a). However, attempts to confirm this theory through manipulations of trunk position were inconclusive (Chapter 3). Alternatively, pelvis position may help to explain age-related differences in hip kinetics and kinematics (Judge et al., 1996b; Kerrigan et al., 1998) and should be explored in future studies.

The findings of this study are limited. In order to control for the effect of sex, and decreased physical activity with age, all participants in this study were male endurance runners which may limit the ability to generalize these findings to women, or to normally active or pathological older adults. The objective function here was squared stress minimization. While this objective function is commonly used for estimating muscle forces in young adults, it is unknown if there is a more appropriate cost function for determining muscle forces in healthy older adults. Squared stress minimization has shown the best agreement with EMG compared to squared muscle force or cubed muscle stress

minimization (Glitsch & Baumann, 1997; Pedotti et al., 1978). Only 43 muscles were included in this model, and only the sagittal and frontal planes were considered when calculating muscle forces. Many of the omitted muscles primarily operate at the hip in planes other than the sagittal plane, and since walking is primarily a sagittal plane activity, this omission likely did not affect muscle forces to a large extent. Additionally, EMG was not used as an input to 'drive' the model. EMG can be difficult to obtain from deep muscles, and is not always reliable and therefore may not be the appropriate choice for calculating muscle forces. However, in the present study, EMG was used to validate timing of predicted muscle forces, and was found to be in good agreement (Fig. 4). Age-related changes in muscle architecture and force-producing properties of muscle were not taken into account in the musculoskeletal model. The older participants in this study reported, on average, 30 years of habitual endurance running. Therefore, age-related differences in these young and older adults may be less than differences reported in the literature, and using these parameters may have overestimated differences in muscles forces between groups. Finally, as this was an exploratory study with many statistical comparisons, critical p was not adjusted for multiple comparisons, increasing the likelihood of false positive results.

5.5 Conclusion

An age-related distal to proximal shift was observed in lower extremity muscle forces such that older adults generated more force in the gluteus maximus and gluteus medius muscles (hip extensors), and less force in the psoas and iliacus muscles (hip flexors) and medial and latera gastrocnemius muscles (plantarflexor). Further, fascicle length of

the medial gastrocnemius was shorter in older adults, indicating decreased force production capabilities consistent with the smaller gastrocnemius muscle force observed in older adults. The presence of these age-related differences in older adults suggest that the primary sources of the age-related distal to proximal shift are, in part, due to age-related differences in muscle architecture and muscle force production. Future research should determine if the observed age-related architectural differences and age-related differences in muscle force are preventable through targeted interventions (e.g. weight training).

Table 5.1. Muscles and corresponding PCSA included in the static optimization procedure.

Muscle	PCSA
Adductor brevis	5.0
Adductor longus	6.5
Adductor magnus	-
Adductor magnus 1	4.3
Adductor magnus 2	5.6
Adductor magnus 3	7.3
Biceps femoris long head	11.3
Biceps femoris short head	5.1
Extensor digitorum longus	5.6
Extensor hallucis longus	2.7
Flexor digitorum longus	4.4
Flexor hallucis longus	6.9
Gastrocnemius medial head	21.1
Gastrocnemius lateral head	9.7
Gemelli	6.4
Gluteus maximus	-
Gluteus maximus 1	10.4
Gluteus maximus 2	13.9
Gluteus maximux 3	8.8
Gluteus medius	-
Gluteus medius 1	14.2
Gluteus medius 2	9.9
Gluteus medius 3	11.3
Gluteus minimus	-
Gluteus minimus 1	3.0
Gluteus minimus 2	3.1
Gluteus minimus 3	3.5
Gracilis	2.2
Iliacus	9.9
Pectineus	7.5
Peroneus brevis	10.4
Peroneus longus	4.9
Peroneus tertius	2.7
Piriformis	8.7
Psoas	7.7
Quadratus femoris	4.2
Rectus femoris	13.5
Sartorius	1.9
Semimembranosus	18.4
Semitendinosus	4.8
Soleus	51.8
Tensor fascia latae	8.2
Tibialis anterior	10.9
Tibialis posterior	14.4
Vastus intermedius	16.4
Vastus lateralis	35.8
Vastus medialis	23.4
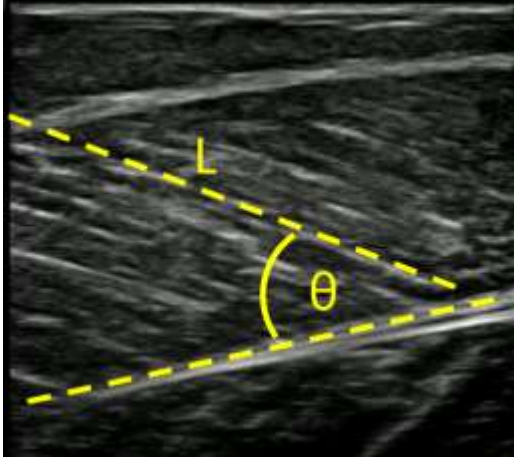
Figure 5.1. Ultrasound image of gastrocnemius medialis. L: fascicle length, θ: pennation angle.
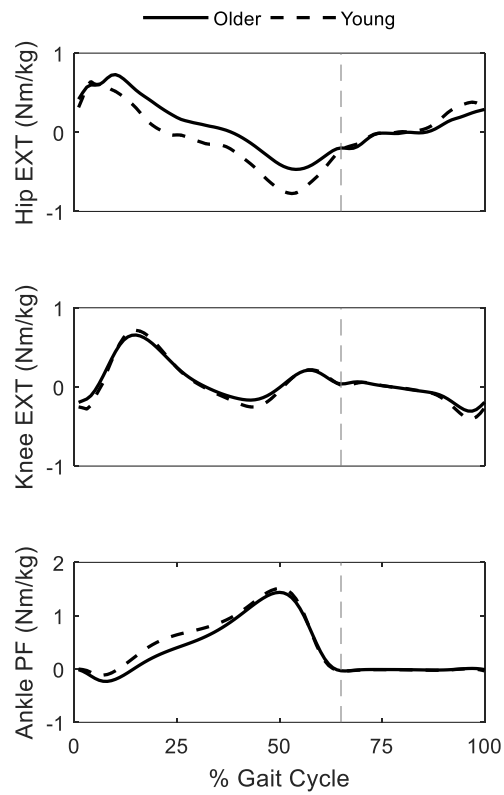
Figure 5.2. Mean sagittal plane joint moments for the hip (top), knee (middle), and ankle (bottom) in young (dashed black) and older (solid black) adults during walking at 1.3 m/s. Positive values indicate extensor torque, negative values indicate flexor torque. The stride begins and ends at heel strike. Vertical dashed line indicates toe-off.
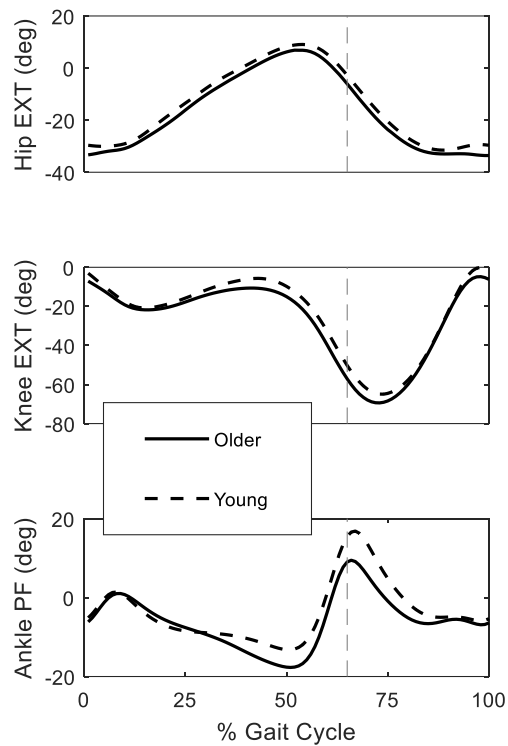
Figure 5.3. Mean sagittal plane joint angular position for the hip (top), knee (middle), and ankle (bottom) in young (dashed black) and older (solid black) adults during walking at 1.3 m/s. Positive values are extension or plantarflexion, negative values are flexion or dorsiflexion. The stride begins and ends at heel strike. Vertical dashed line indicates toe-off.
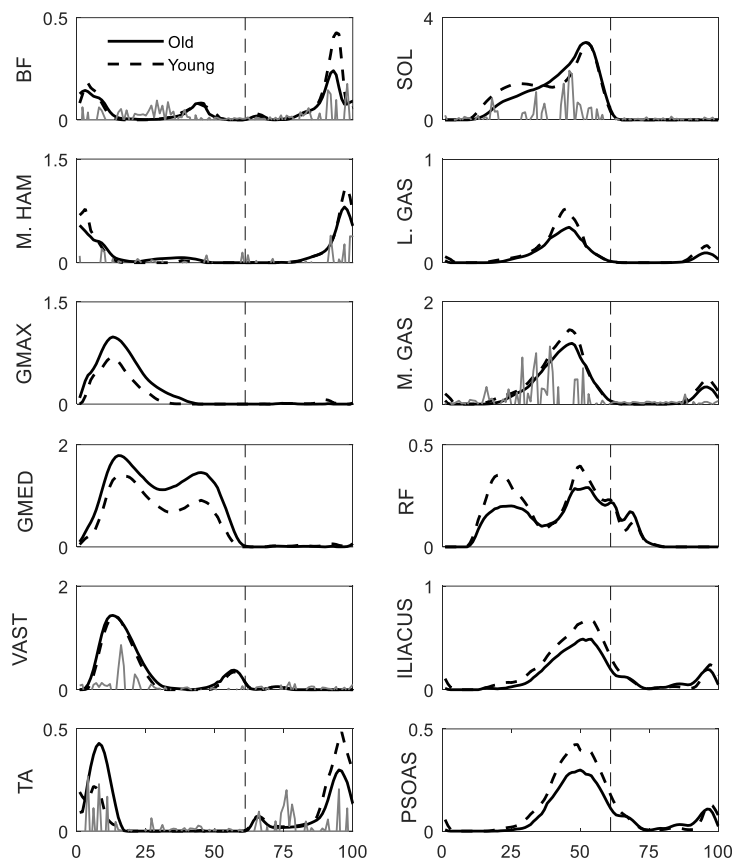
Figure 5.4. Mean muscle force in young (dashed black) and older adults (solid black). Units are BW. Muscles shown are: biceps femoris (BF), medial hamstrings (M. HAM), gluteus maximus (GMAX), gluteus medius (GMED), vasti (VAST), tibialis anterior (TA), soleus (SOL), lateral gastrocnemius (L. GAS), medial gastrocnemius (M. GAS), rectus femoris (RF), iliacus (ILIACUS), and psoas(PSOAS). EMG for one representative older adult subject is shown in gray. EMG was arbitrarily scaled.
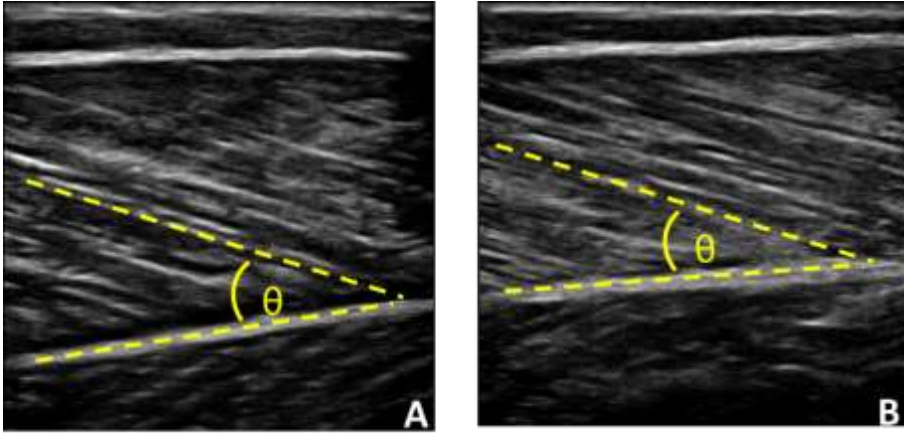
Figure 5.5. Representative ultrasound images of the medial gastrocnemius from A) young adult man, and B) older adult man, with fascicle pennation angle (θ) illustrated.

CHAPTER 6

CONCLUSION

6.1 Summary

The overall objective of this dissertation was to investigate potential sources of the age-related distal to proximal shift in lower extremity joint kinetics. While this shift has been well-documented (Cofre et al., 2011; DeVita & Hortobagyi, 2000a; Kerrigan et al., 1998; Winter et al., 1990), the specific sources responsible for the shift are not well-understood. Therefore, the studies that comprise this dissertation were designed to determine the extent to which the age-related distal to proximal shift in joint kinetics could be explained by age-related differences in (i) step length and trunk kinematics, (ii) habitual physical activity, in the form of endurance running, and (iii) gastrocnemius muscle architecture and lower-extremity muscle forces. Chapter 3 investigated the easily modifiable factors of step length and trunk position, which have been speculated to influence age-related differences in gait kinetics but, prior to this dissertation, had not yet been systematically tested. Chapter 4 investigated a less easily modifiable parameter, years of endurance running, by comparing older adult endurance runners and non-runners to determine the effect of habitual physical activity on the age-related distal to proximal shift. Finally, Chapter 5 investigated the age-related distal to proximal shift in kinetics at the muscle level by comparing age-related differences in muscle architecture and estimated muscle forces.

Chapter 1 of this document presented three formal hypotheses for Chapters 3 and 4. No formal hypotheses were presented for Chapter 5 (due to its exploratory nature and

large number of variables). The hypotheses, results and conclusions of each study were addressed in their respective chapters, and are revisited here.

A Note about Physical Activity

First, it is important to make clear that the 'physically active' participants in these studies were limited to endurance runners. Therefore, conclusions made in Chapter 4 about the effect of physical activity are limited to older adult endurance runners and do not necessarily generalize to older adults who engage in habitual strength training or who engage in other forms of endurance exercise (e.g. cycling), or those who are just highly active in their daily lives but do not often perform bouts of "exercise" as we usually define that term. Relatedly, one novel aspect of this work is the use of highly trained young and older endurance athletes in Chapters 3 and 5, studies in which physical activity level was not an independent variable. The effect of age-related differences in physical activity on age-related differences in gait kinetics and muscle architecture is unclear (Boyer et al., 2012; Buddhadev & Martin, 2016; Savelberg et al., 2007). Therefore, using highly physically active older adults (i.e. endurance runners) to address questions pertaining to gait kinetics and muscle architecture provided a means of accounting for the potential confounding relationship of physical activity and allowed for interpretation of differences based on the effect of age.

Chapter 3

Hypothesis 1.1 stated that older adults would exhibit smaller hip kinetics and larger ankle kinetics when walking with a step length similar to young adults, compared to walking with a self-selected step length, and hypothesis 1.2 stated that older adults would exhibit smaller hip kinetics when walking with a more upright trunk angle compared to a self-selected trunk angle. Older adult participants in this study self-selected a similar step length compared to young adults when walking at 1.3 m/s and therefore did not display differences in kinetics when deliberately walking with the same step length as young adults at 1.3 m/s, thus hypothesis 1.1 was not supported. Additionally, no differences in kinetics were observed when older adults deliberately walked with a more upright trunk compared to walking with a self-selected trunk position, thus hypothesis 1.2 was also not supported. In summary, even when older adults had similar preferred step lengths and walked with a more upright trunk posture, the age-related distal to proximal shift of joint kinetics was not reversed or substantially mitigated.

It is possible that some untested condition, e.g. a different set of step lengths, trunk postures, and/or walking speed may elicit age-related differences in kinetics, but given the present results, these kinematics would likely be quite different from the typical kinematics of healthy young adults, and the utility of such results would be debatable. Relatedly, the lack of differences in hip kinetics or kinematics when older adults were tasked with walking more upright, despite obvious differences in trunk lean, suggest that pelvis position may play an important role in age-related differences in hip mechanics.

Chapter 4

Hypothesis 2.1 stated that older non-runners would display smaller ankle kinetics and larger hip kinetics compared to older runners. The older runners in this study reported, on average, a history of 30+ years of running and were currently running at least 20 miles a week. Despite high levels of running related fitness and greater isometric hip extensor, knee extensor, and dorsiflexor strength compared to older non-runners, non-runners did not display smaller ankle kinetics or larger hip kinetics compared to runners. In effect, high levels of running did not reverse or reduce the age-related distal to proximal shift of joint kinetics, thus hypothesis 2.1 was not supported. These results suggest that, in otherwise healthy older adults with a high level of running related fitness, the characteristic age-related differences in joint kinetics are not modifiable and are an unavoidable part of natural aging.

The similar plantarflexor 'strength' (i.e. peak isometric torque) between older runners and non-runners suggests that 1) although running may promote greater hip and knee extensor strength, it is not sufficient for promoting greater plantarflexion strength in older adults, and 2) due to the relationship between maximum isometric plantarflexor torque and peak ankle power during walking, the similar plantarflexor strength contributed to the similar ankle kinetics between older runners and non-runners. It is unknown whether an alternative mode of exercise (e.g. strength training) would be effective in reversing or reducing the age-related distal to proximal shift.

Chapter 5

Due to the exploratory nature of this study and the large number of variables, formal hypotheses were not established, but it was generally expected that older adults would display larger hip extensor muscle forces and smaller plantarflexor muscle forces compared to young adults, and young and older adults would display differences in medial gastrocnemius fascicle length and pennation angle. Older adults generated larger peak forces in the gluteus maximus and gluteus medius muscles (hip extensors), smaller peak forces in the iliacus and psoas muscles (hip flexors), smaller peak forces in the medial and lateral gastrocnemius muscles (ankle plantarflexors), and no difference in peak soleus force (ankle plantaflexor), compared to young adults. The age-related difference in gastrocnemius force and lack of difference in soleus force may be partially due to static optimization techniques in which assignment of force is preferential to muscles with large PCSA's, such as the soleus. However, it is also likely that the smaller gastrocnemius muscle forces are due to the biarticularity of the gastrocnemius. Since the gastrocnemius also crosses the knee joint, the greater knee flexion in old vs. young adults may have resulted in smaller gastrocnemius forces in older vs. young adults and should be explored further in future studies.

Medial gastrocnemius pennation angle was not different between young and older adults, which may be due to the present subjects' history of endurance running (Karamanidis & Arampatzis, 2005) and thus greater pennation angle. However, fascicle length was shorter in older adults compared to young, indicating decreased force production capabilities in the gastrocnemius consistent with the smaller gastrocnemius muscle force observed in older adults. Summarily, these age-related differences in muscle

forces and muscle architecture indicate 1) the distal to proximal shift is present at the muscle level, and smaller ankle moments, may be influenced specifically by smaller gastrocnemius muscle forces and 3) the gastrocnemius may be the primary site of age-related adaptations contributing to differences in ankle kinetics between young and older adults.

6.2 General Conclusions

The findings of this dissertation support the notion that the age-related distal to proximal shift of kinetics in healthy older adults is due primarily to age-related differences at the muscle level and do not support previous speculations that these differences are due to spatiotemporal factors such as step length (Judge et al., 1996a), joint kinematics (Judge et al., 1996b; Kerrigan et al., 1998) or physical activity (Boyer et al., 2012). These findings are particularly noteworthy for two reasons: 1) habitual physical activity, in the form of years of endurance running, is not sufficient for reversing or reducing the age-related shift in kinetics, and may suggest this shift is a largely unavoidable part of healthy aging, and 2) the smaller plantarflexor moments in older vs young adults may be primarily due to smaller muscle forces in the gastrocnemius muscles and not the soleus muscle. This finding may be clinically meaningful for future research aiming to develop targeted training programs to increase or preserve mobility in older adults.

6.3 Future Work

The present interpretations about the effect of physical activity are limited to older adults who engage in years of endurance running, therefore future work should characterize the effects of varying modes of physical activity (i.e. strength training vs. endurance training) on kinetics in older adults. Relatedly, since strength training and endurance training may result in different adaptations to muscle architecture (Blazevich, 2006), age-related differences in muscle architecture and muscle forces should be explored in older adults who engage in habitual strength training. Lastly, future work should explore the clinical implications of observed age-related differences in muscle architecture and muscle forces. For example, follow up studies can be designed to determine if physical training in sedentary older adults can reverse or reduce the observed age-related architectural differences.

APPENDIX A

PARTICIPANT PHYSICAL ACTIVITY DETAILS

Older adult endurance runners:

Participant ID	Est. Running History	Current Training
01	20 yrs	20-25 miles/wk
02	40 yrs	35-40 miles/wk
03	35 yrs	35 miles/wk, calisthenics 2x/wk
04	50 yrs	20-30 miles/wk, resistance training 2x/wk
05	60 yrs	25-30 miles/wk, resistance training 3x/wk
06	45 yrs	25-30 miles/wk, resistance training 4x/wk
07	35 yrs	20 miles/wk, resistance training 2x/wk, swim 2x/wk
08	35 yrs	25 miles/wk, calisthenics 3x/wk
09	7 yrs	20 miles/wk, aerobics class 3x/wk
10	30 yrs	30 miles/wk
11	40 yrs	30-35 miles/wk
12	15 yrs	20-25 miles/wk, resistance training 2x/wk

Young adult endurance runners:

Participant ID	Current Training
01	30 miles/wk
02	20-30 miles/wk, resistance training 3-4x/wk
03	25 miles/wk, resistance training/calisthenics 4x/wk
04	25-30 miles/wk, resistance training 2x/wk
05	20 miles/wk
06	40-60 miles/wk, calisthenics 3x/wk
07	20-25 miles/wk
08	50 miles/wk
09	40-50 miles/wk, resistance training 2x/wk
10	30-40 miles/wk
11	35-45 miles/wk, resistance training 2x/wk, swim 2x/wk
12	20 miles/wk
13	30 miles/wk, resistance training 4x/wk

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