KINEMATIC AND KINETIC ANALYSIS OF WALKING AND RUNNING ACROSS SPEEDS AND TRANSITIONS BETWEEN LOCOMOTION STATES

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DISSERTATION ABSTRACT

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Title: Kinematic and Kinetic Analysis of Walking and Running across Speeds and Transitions between Locomotion States

Walking and running are general locomotion activities for human beings. Basic gait patterns and whole body center of mass (COM) dynamic patterns are distinctly different between them. Lower extremity joint mechanics patterns could reflect musculoskeletal coordination characteristics. Change of locomotion tasks and speeds can affect lower extremity joint kinematic and kinetic characteristics, and progression of age may also affect these characteristics. Little is known about change of locomotion tasks and speeds effects on lower extremity joint level kinetic characteristics, and whether there is a connection between COM system and lower extremity system. To address this, twenty healthy subjects were recruited to participate in a series of treadmill tests, including walking (0.8 – 2.0 m/s, with 0.2 m/s intervals), running (1.8 – 3.8 m/s, with 0.4 m/s intervals) and gait mode transition from walking to running, and from running to walking (between 1.8 – 2.4 m/s, $\pm$ 0.1 m/s^2). Three-dimensional kinematic and kinetic data were collected in all locomotion tests and used to calculate and analyze outcome variables for lower extremity joints and the COM system across different conditions. Results indicate that change of locomotion speeds significantly affect joint level kinetic

characteristics within both walking and running locomotion states. Different locomotion task demands (walking vs. running) require fundamental alteration of lower extremity joint level kinetic patterns, even at the same locomotion speed. Progression of age also affects lower extremity joint level kinematic and kinetic patterns in walking and running across speeds. Additionally, stance phase an energy generation and transfer phenomenon occurred between the distal and proximal joints of the lower extremity in both walk-to-run and run-to-walk transitions. Lastly, a connection exists between whole body COM oscillation patterns and lower extremity joint level kinetic characteristics in running. These findings serve to further clarify the mechanisms involved in change of locomotion tasks and speeds effects on lower extremity joint kinetic patterns, and further establish a connection between the COM system and the lower extremity system. These findings may be beneficial for future foot-ankle assistive device development, potential optimization of gait efficiency and performance enhancement.

This dissertation includes previously published and unpublished coauthored material.

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CHAPTER I

INTRODUCTION

Background and Significance

As part of common daily life activities, locomotion plays an important role for human beings. Walking and running are two general forms of locomotion, each having their own unique patterns in ground reaction force (GRF) (G. A. Cavagna, Saibene, & Margaria, 1964; G. Cavagna, Saibene, & Margaria, 1963; Cavanagh & Lafortune, 1980; Elftman, 1939, 1940; C T Farley & Ferris, 1998), general limb support patterns, and whole body center of mass (COM) mechanical energy fluctuation characteristics (G. A. Cavagna & Margaria, 1966; G. A. Cavagna et al., 1964; G. A. Cavagna, Thys, & Zamboni, 1976; G. Cavagna et al., 1963; C T Farley & Ferris, 1998). Specifically, walking is featured with at least one foot in contact with ground and a small period of double limb support (C T Farley & Ferris, 1998); whereas running is characterized with only one leg in contact with ground and lower extremity musculoskeletal system is relatively compliant compared with walking (C T Farley & Ferris, 1998). Walking can generally be described as a “vaulting” movement over relatively stiff legs (Geyer, Seyfarth, & Blickhan, 2006; Jin & Hahn, 2018; McGowan, Grabowski, McDermott, Herr, & Kram, 2012), while running can be described as a “bouncing” movement over relatively compliant legs (Geyer et al., 2006; Jin & Hahn, 2018; McGowan et al., 2012). These substantial differences result in subsequent COM dynamic pattern differences between walking and running: specifically, COM gravitational potential energy (E_{pot}) and mechanical kinetic energy (E_{kin}) are out-of-phase in walking, but are in-phase during running (Veerle Segers, Aerts, Lenoir, De Clercq, & De Clerq, 2007). This indicates that there is a large amount of pendulum-type energy exchange between E_{pot} and E_{kin} in walking (G. A. Cavagna & Margaria, 1966; G. Cavagna et al., 1963; C T Farley & Ferris, 1998), while not much energy transfer occurs between them in running (G. A. Cavagna et al., 1964; C T Farley & Ferris, 1998). These differences indicate that walking and running pose different locomotion task demands to human beings. Different gait strategies would be needed to meet these demands. Additionally, within walking or running locomotion state, change of speeds would also pose different requirements on the body, associated with potential gait kinematic and kinetic patterns change.

The lower extremity musculoskeletal system plays an important role in both walking and running activities across speeds, as well as in changes of locomotion task demand. This involves body support during stance phase, maintenance of whole body COM dynamic stability, energy absorption and generation in both stance and swing phase to assist with forward movement, etc. Moreover, maintaining dynamic balance and optimizing gait efficiency among different speeds within the same locomotion task, or switching gait patterns when locomotion task demand changes, are also important functional requirements for lower extremity musculoskeletal system.

If we assumed the lower extremity to be a dynamic organization system, each movement would arise from coordination and work of numerous neuromuscular and musculotendinous systems (C T Farley & Ferris, 1998). Based on the theory that within a system which is composed of higher and lower levels of organization, lower level organizations would have more complex components (C T Farley & Ferris, 1998). A good way to investigate lower extremity musculoskeletal system characteristics in locomotion activities could start from relatively higher level organization within lower extremity system (C T Farley & Ferris, 1998): joint level kinematic and especially joint kinetic patterns.

Lower extremity joint level kinematic and kinetic characteristics can reflect musculoskeletal system function in locomotion activities. And joint level elasticity, mechanical work and power absorption and generation are important characteristics associated with musculoskeletal system dynamic function in different locomotion speeds (Anahid Ebrahimi, Goldberg, & Stanhope, 2017; G. A. Cavagna, 1977; Kuitunen, Komi, Kyröläinen, & Kyrolainen, 2002). Joint elasticity, also known as joint torsional stiffness or dynamic stiffness (K_{joint}), reflects sagittal plane joint spring-like stretching and shortening characteristics under loading conditions (Kuitunen et al., 2002). Joint mechanical work (W_{joint}) and power (P_{joint}) indicate that the musculotendinous system receive and release mechanical energy and rate of work performed in a stretch-shorten cycle (G. A. Cavagna & Kaneko, 1977).

Previous studies have investigated the effect of speeds change on K_{joint}, W_{joint} and P_{joint} in both walking and running. For K_{joint} investigation in walking, the ankle, knee and hip joint angle-moment curves, slopes have been observed to increase with walking speeds (Frigo, Crenna, & Jensen, 1996). In running, it was reported that K_{ankle} would remain relatively consistent from slow to sprinting speeds (Arampatzis, Bruk, & Metzler, 1999; Kuitunen et al., 2002). However, other research reported that K_{ankle} was higher at sprinting speeds compared with slow speed running (Stefanyshyn & Nigg, 1998). Additionally, K_{knee} has been reported to increase with running speeds (Arampatzis et al., 1999; Kuitunen et al., 2002). Based on these findings, it seems that previous studies did not include all lower extremity K_{joint} (K_{ankle}, K_{knee} and K_{hip}) into the analysis, and it remains that little is known about whether change of locomotion task demand would have influence on K_{joint} patterns. Previous findings for W_{joint} and P_{joint} were mixed: it was reported that when walking and running speeds changed, ankle, knee and hip joint's relative power contribution to total power did not change (Farris & Sawicki, 2012), and P_{joint} did not increase from walking to sprinting (Schache, Brown, & Pandy, 2015); while another study reported stance phase positive and negative W_{ankle} increased with speeds (Anahid Ebrahimi et al., 2017). With these mixed findings, more details need to be investigated about positive and negative W_{joint} and P_{joint} patterns in stance and swing phase in walking and running. This would provide more details about each joint's functional role and mechanical characteristics in different phases of walking and running across a range of speeds. Moreover, previous studies investigated joint stiffness, joint mechanical work and power separately. Little is known about the relationship between stance phase joint dynamic loading and response characteristics, and joint energy absorption and generation patterns across speeds within walking and running activities. Based on these unknown and mixed findings, it is necessary to further investigate the effects of locomotion speeds and task demand effects on lower extremity K_{joint}, W_{joint} and P_{joint} characteristics.

Human locomotion activities can be affected and constrained by physiological and biomechanical factors (Chung & Wang, 2010). The ability to maintain dynamic balance and gait efficiency is a fundamental skill for human beings in locomotion activities (Shkuratova, Morris, & Huxham, 2004). However, balance control capabilities have been reported to be compromised with the progression of age (Shkuratova et al., 2004). Muscle strength is another important factor which is known to be associated with locomotion performance (Akbari & Mousavikhatir, 2012). The degeneration of balance control and muscle strength capabilities associated with increased age, can then lead to reduction of functional performance (Akbari & Mousavikhatir, 2012; Bottaro, Machado, Nogueira, Scales, & Veloso, 2007; Melzer, Kurz, & Oddsson, 2010; Wang, Olson, & Protas, 2002), and subsequently influence the general walking and running gait characteristics.

Previous studies have found that as age increased, self-selected locomotion speed and step length would decrease (Judge, Davis, & Ounpuu, 1996; Kerrigan, Todd, Della Croce, Lipsitz, & Collins, 1998; Silder, Heiderscheit, & Thelen, 2008; Winter, Patla, Frank, & Walt, 1990). To walk or run at the same locomotion speed and to maintain similar dynamic balance and gait efficiency compared with young age people, different movement strategies and musculoskeletal coordination patterns would be needed for middle-age and elderly individuals. This may in turn require some compensatory mechanisms among lower extremity joints and segments, which can be measured as joint level kinematic and kinetic characteristics. Based on previous studies about age effect on joint level kinematic and kinetic characteristics in walking and running, it has been reported that elderly people tended to increase hip joint positive work to compensate for decreased work generated at the ankle joint in walking (Browne & Franz, 2017; DeVita, Hortobagyi, & Carolina, 2000; Silder et al., 2008). In running gait, it has been reported that knee joint range of motion in sagittal plane, and the associated shock-absorbing capacity was decreased in elderly people (Bus, 2003; Fukuchi & Duarte, 2008). This may indicate K_{knee} in running would tend to increase with age. Most of these findings were focused on comparisons between young and older adults at self-selected locomotion speeds. However, little is known about whether a smaller range of age differences would affect lower extremity joint kinematic and kinetic characteristics. Further investigation about whether lower extremity joint kinematic and kinetic patterns are different between young and middle age healthy people across different locomotion tasks and speeds is needed.

Along with investigation about age effects on change of gait kinematic and kinetic patterns, the relationship between change of locomotion speeds and switch between locomotion states (gait transition) also warrants further investigation. When walking at a constantly increasing speed, or running at a constantly decreasing speed, spontaneous walk-to-run transition (WRT) or run-to-walk transition (RWT) has been observed to occur at preferred gait transition speeds (PTS) (Raynor, Yi, Abernethy, & Jong, 2002). Humans tend to optimize metabolic and mechanical efficiency during locomotion, which subsequently affects dynamic function and gait performance. Considering this, it follows that walking at speeds higher than PTS, or running at speeds lower than PTS would not be optimal for locomotion efficiency.

There are many factors which help explain what triggers gait transition between walking and running. Generally, there have been four proposed mechanisms which modulate and trigger gait transition: metabolic efficiency, mechanical efficiency, mechanical load, and cognitive and perceptual modulation (Kung, Fink, Legg, Ali, & Shultz, 2018). Among these proposed mechanisms, mechanical efficiency and mechanical load mechanisms are the important factors contributing to gait transition from a biomechanical perspective (Kung et al., 2018; Pires, Lay, & Rubenson, 2014). Lower extremity joint level kinetic patterns are associated with gait transition triggers:

musculoskeletal system mechanical efficiency and mechanical load optimization mechanisms. Previous studies have reported that when walking at speeds above PTS, more effort is required from ankle and hip muscles (Pires et al., 2014). Comparing walking and running at PTS, lower extremity joint power generation has been reported to shift from proximal to distal joints in running (Farris & Sawicki, 2012). This indicates that gait transition from walking to running is beneficial for positive mechanical energy generation (Farris & Sawicki, 2012). Moreover, in a study focused on the WRT process, lower extremity joint moment and power characteristics at the transition step were reported to be similar to running gait (V. Segers, De Smet, Van Caekenberghe, Aerts, & De Clercq, 2013). Previous studies have focused on investigation of walking above PTS or running below PTS to explore possible mechanisms which trigger gait transition, as well as WRT process gait kinematic and kinetic patterns. However, little is known about stance phase joint dynamic loading and response, or joint kinetic characteristics in both stance and swing phase throughout the gait transition process. Further investigation of stance phase joint level stiffness, joint extensor moment angular impulse, as well as stance and swing phase joint kinetics (work and power) is necessary.

Switching of gait patterns (gait transition) in response to speed change reveals that walking and running general gait characteristics are different. Previous statements have been focused on investigation on change of locomotion tasks and speeds, as well as progression of age potential effects on lower extremity joint level characteristics. Human whole body COM dynamic movement patterns are closely related to lower extremity joint mechanical characteristics. Examination of COM dynamic characteristics can not only provide a connection with lower extremity kinetic patterns, but also give a bigger picture of the whole body dynamic system's kinetic function as well. Little is known about the relationship between lower extremity joint mechanics patterns and whole body center of mass dynamic characteristics across a range of speeds. Further investigation of the relationship and coordination characteristics is necessary. Additionally, since musculoskeletal system stiffness was reported to be associated with performance (Brughelli & Cronin, 2008; Lindstedt, Reich, Keim, & LaStayo, 2002; Reich, Lindstedt, LaStayo, & Pierotti, 2000), it would be necessary to further investigate joint level stiffness and the higher level whole leg stiffness (K_{leg}) across different speeds. This investigation would lay a better knowledge framework of improving gait performance.

Based on previously mentioned walking and running general gait patterns, lower extremity system stiffness and COM dynamic oscillation patterns, a simplified inverted pendulum model (Adamczyk & Kuo, 2009; Kuo, 2007; McGrath, Howard, & Baker, 2015) and spring-mass model (Alexander, 1992; C T Farley, Glasheen, & McMahon, 1993; Claire T Farley & Gonzalez, 1996; McGowan et al., 2012; McMahon & Cheng, 1990) have been proposed and used in walking and running gait analysis, respectively. Due to the different assumptions of these two models, only vertical stiffness (K_{vert}), leg stiffness (K_{leg}), and joint stiffness (K_{joint}) could be calculated and derived from the spring-mass model in running (Brughelli & Cronin, 2008). To further investigate the relationship of K_{joint} and K_{vert}, K_{leg} across different speeds, selection of the general running condition is realistic and reasonable. K_{vert} reflects COM oscillation characteristics in running stance phase (Brughelli & Cronin, 2008; G. Cavagna, Franzetti, Heglund, & Willems, 1988; McMahon, Valiant, & Frederick, 1987), while K_{leg} reflects the ground contact period leg length change connection between the foot and the COM

(Brughelli & Cronin, 2008; C T Farley et al., 1993; McMahon & Cheng, 1990). Previous studies have found K_{vert} tended to increase with running speeds (Brughelli & Cronin, 2008; G. A. Cavagna, 2005; He, Kram, & McMahon, 1991; Morin, Dalleau, Kyröläinen, Jeannin, & Belli, 2005) while K_{leg} remained relatively unchanged (Biewener, 1989; Brughelli & Cronin, 2008; C T Farley et al., 1993; He et al., 1991; McMahon & Cheng, 1990; Morin et al., 2005; Morin, Jeannin, Chevallier, & Belli, 2006). Within the lower extremity musculoskeletal system, since leg spring system stiffness (K_{leg}) characteristics may emerge from local K_{joint} adjustment (C T Farley, Houdijk, Van Strien, & Louie, 1998; Claire T. Farley & Morgenroth, 1999; Günther & Blickhan, 2002; Sholukha, Gunther, & Blickhan, 1999), further investigation about whether K_{vert} and K_{leg} could be predicted from K_{joint} in different running speeds is necessary.

Additionally, investigation of COM mechanical patterns (work and power) would be beneficial for a better understanding of gait performance enhancement. Previous studies have investigated COM mechanical work (W_{com}) and instantaneous power (P_{com}) in walking (Adamczyk & Kuo, 2009; Donelan, Kram, & Kuo, 2002; Zelik & Kuo, 2010) and WRT steps (Veerle Segers et al., 2007). However, little is known about W_{com} and P_{com} characteristics in running across different speeds, and the relationship between W_{com} and K_{vert}, K_{leg} respectively across speeds. Further investigations of the COM dynamic and mechanical characteristics, as well as the relationship with lower extremity system kinetic patterns are needed.

General and Specific Aims

The overall goal of this study was to investigate whether change of locomotion speeds or tasks would have influence on lower extremity joint level kinematic and kinetic characteristics; and lower extremity joint level kinetic patterns during the gait transition (WRT, RWT) process. A second goal was to investigate whole body COM oscillation patterns and mechanical characteristics in running across speeds, and the relationship between COM dynamic characteristics and lower extremity kinetic patterns. The anticipated outcomes of this project would be beneficial for assistive device development and gait performance enhancement. First, findings from change of locomotion tasks and speeds effects on joint level kinetic characteristics (stiffness, work and power) would be beneficial for future assistive device development, which aims to have adjustable stiffness and work performed characteristics in response to speeds and gait patterns change. Second, knowledge of the gait transition process and compensatory mechanisms within lower extremity joints would be beneficial for understanding the functional role of each joint during the transition process. Finally, this dissertation seeks to find the relationship between whole body COM dynamic characteristics and lower extremity stiffness patterns, laying a better knowledge framework for gait performance enhancement. The goals of this dissertation will be addressed through four specific aims.

Specific Aim 1. To investigate change of locomotion tasks (walking vs. running) and speeds (within walking and running state) effects on lower extremity joint level stiffness, stance and swing phase mechanical work and average power characteristics. It was hypothesized that: (1) lower extremity joint stiffness would increase when locomotion speeds increased; (2) joint stiffness, joint work and power would be higher in running compared with walking.

Specific Aim 2. To investigate lower extremity joint level kinematic and kinetic characteristics, general gait patterns between young and middle age group in both walking and running, across speeds. We also sought to identify whether there is a compensatory mechanism among lower extremity joints in middle aged adults in a wide range of walking and running speeds. It was hypothesized that the middle age group would have: (1) higher joint stiffness; (2) higher stance phase hip joint extensor moment angular impulse and positive work, lower ankle joint plantar flexor moment angular impulse and positive work; and (3) smaller joint angle range of motion, step length and higher gait cadence compared with a young age group.

Specific Aim 3. To investigate lower extremity joint stiffness, stance phase joint extensor moment angular impulse, stance and swing phase joint work and power characteristics in the WRT and RWT process. It was hypothesized that: (1) lower extremity joint stiffness would increase during WRT, and decrease in the RWT process; (2) joint work, peak power and extensor moment angular impulse would increase during the WRT, and decrease in the RWT process.

Specific Aim 4. To investigate whether change of running speeds would have influence on change of K_{vert} and K_{leg} patterns, and whole body W_{com} and P_{com} characteristics. Another goal was to investigate whether K_{vert}, K_{leg} can be predicted from K_{joint} within each running speed. Moreover, we also planned to investigate whether a connection occurred between sagittal plane W_{coms}^{+} and K_{vert}, K_{leg} respectively across running speeds. It was hypothesized that: (1) K_{vert}, W_{com} would increase with running speeds while K_{leg} would remain relatively unchanged; (2) K_{vert} and K_{leg} would be predicted more from K_{knee}, compared with K_{ankle} and K_{hip} at each speed; (3) W_{com} could be predicted from K_{vert} and K_{leg} across running speeds.

Organization of Dissertation

This dissertation is written in a journal format style, where Chapters III-VI have been or will be submitted for publication to peer-reviewed journals. The following explains how these chapters fit together into a coherent body of work. A bridge paragraph is presented at the end of Chapters III-V to provide context to flow from one chapter to the next.

The current chapter (Chapter I) has provided the background information and significance necessary to detail how the research questions of this dissertation were formulated and has described the general and specific aims that have guided the overall study. Next, Chapter II will detail the methodology implemented for each study, while explaining the similarities and differences between each. Chapter III will compare general change of locomotion tasks and speeds effect on lower extremity joint kinetic patterns among young healthy subjects. Chapter IV will add middle age group subjects' data to compare effects of age (young vs. middle age) on lower extremity joint kinematic and kinetic characteristics. Chapter V will examine the change of locomotion state process (gait transition) between walking and running (WRT, RWT) among middle age subjects, further expanding the knowledge framework investigated in Chapter III-IV. Chapter VI incorporates the findings from Chapter III-V about lower extremity joint level kinematic and kinetic characteristics, and delves deeper to investigate the connection between whole body COM dynamic patterns and lower extremity system kinetic characteristics. Finally, Chapter VII provides a summary of the key findings from the overall body of work, giving a larger-picture view of this set of studies while mentioning limitations and suggesting directions for future work.

This dissertation includes co-authored work, some of which has already been published in peer-reviewed journals. Chapter III of the dissertation has already been published in *Human Movement Science*. Chapter IV-VI of the dissertation will be submitted for publication soon to appropriate journals. For all work in this dissertation, Li Jin was the primary contributor, including being responsible for designing the study, subject recruitment, data collection, data analysis, and dissemination. Michael E. Hahn, the other co-author on this set of studies, oversaw all aspects of the dissertation process from a mentorship role and participated in study design as well.

CHAPTER II

GENERAL METHODOLOGY

Subjects

To address Specific Aim 1 (Chapter III), ten young healthy subjects (5 males, 5 females; 23 $\pm$ 5.3 years, 170 $\pm$ 11.2 cm, 67 $\pm$ 14.2 kg) were recruited. To address Specific Aim 2 (Chapter IV), another ten middle-age healthy subjects (5 males, 5 females; 51 $\pm$ 6.0 years, 173 $\pm$ 11.4 cm, 70 $\pm$ 15.0 kg) were added to the data set in Chapter III and these two sets of data were analyzed together, to compare lower extremity joints kinematic and kinetic characteristics between young-age group and middle-age groups. To address Specific Aim 3 (Chapter V), the ten middle-age healthy subjects from Chapter IV were selected for the investigation of lower extremity joint kinetic patterns in gait transition process. To address Specific Aim 4 (Chapter VI), both young-age group and middle-age groups were combined and analyzed to investigate COM dynamic patterns.

The age range selection criteria for the young-age group was between 18 – 35 years old, and for the middle-age group was between 40 – 60 years old. Subjects were excluded based on any of the following criteria: a history of neurologic deficits or other musculoskeletal disorders that would affect gait, a history of rheumatic diseases, or a history of unexpected falls in the previous six months. For Chapters III-VI, informed consent was obtained from subjects, and all study protocols were approved by the University of Oregon Institutional Review Board.

Study Design and Experimental Protocol

Chapter III, IV, VI

Subjects were first instructed to walk on a force-instrumented treadmill (Bertec, Inc., Columbus, OH) at seven increasing speeds, from 0.8 to 2.0 m/s (at 0.2 m/s intervals), for 90 seconds per stage. Then they were asked to run at six different speeds, from 1.8 to 3.8 m/s (at 0.4 m/s intervals), for 75 seconds per stage. Walking conditions were tested before running conditions, and there was a break between walking and running conditions. Subjects were allowed to rest at any time during the testing. Only the running protocol was selected for data analysis in Chapter VI.

Chapter V

Subjects were first asked to complete the WRT protocol: walking on a force-instrumented treadmill (Bertec, Inc., Columbus, OH) at 1.8 m/s for 30 seconds, then the treadmill was constantly accelerated at 0.1 m/s^2 up to 2.4 m/s. Subjects were asked to transition to a running gait whenever they felt ready during the acceleration process. After transitioning to a running gait, they ran at 2.4 m/s for another 30 seconds. Next, subjects completed the RWT protocol: running at 2.4 m/s for 30 seconds, then the treadmill was constantly decelerated at -0.1 m/s^2 down to 1.8 m/s. Subjects were asked to transition to a walking gait whenever they felt ready during the deceleration process. Once they transitioned to a walking gait, the subjects walked at 1.8 m/s for another 30 seconds. Treadmill acceleration and deceleration magnitude for the WRT and RWT protocols were chosen based on previous work (V. Segers, Aerts, Lenoir, & De Clercq, 2006).

Data Collection

Chapter III-VI

We measured subjects' body mass, height and leg length (L_0) before the formal test. Leg length (L_0) was measured as the vertical distance from the greater trochanter to the floor during static standing, based on a previously published protocol (McGowan et al., 2012). Fifty-five retro-reflective markers were placed on the skin surface, adapted from a previously published whole body marker set (Sawers & Hahn, 2012). Three-dimensional segmental kinematic data were collected at 120 Hz using an 8-camera motion capture system (Motion Analysis Corp., Santa Rosa, CA). Ground reaction force data were collected at 1200 Hz using the force-instrumented treadmill. Kinematic and kinetic data were filtered with a low-pass fourth-order Butterworth filter at 6 Hz and 50 Hz, respectively. In Chapter III-IV and Chapter VI, data were extracted from the middle strides (20 strides on average) of each stage in walking and running conditions, respectively.

Data and Statistical Analysis

Chapter III

To investigate change of speed and locomotion task effects on lower extremity joint kinetic patterns, lower extremity joint angles, moments and net powers were calculated using an inverse dynamics model coded in Visual 3D (C-Motion, Inc., Germantown, MD). Joint stiffness (K_{joint}), stance and swing phase joint positive work (W^{+}_{joint}) and negative work (W^{-}_{joint}), stance and swing phase joint average positive power (P^{+}_{joint}) and negative power (P^{-}_{joint}) were calculated for ankle, knee and hip joints, respectively. A more in-depth explanation of the variables selection and calculation details is given in Chapter III. All the outcome variables were calculated and averaged from both limbs, normalized to body mass (where appropriate) and averaged across 3 gait cycles.

Statistical analysis was performed using two 2-way ANOVAs (joint × speed); one for walking and one for running conditions**,** in SPSS (V22.0, IBM, Armonk, NY). Initial alpha level was set to 0.05. Bonferroni adjustments were used for pairwise comparisons. When main effect or interaction effect were detected, post hoc analyses were conducted. Follow up pairwise comparison alpha level was set to 0.05 divided by the number of comparisons. A paired sample t-test was conducted to test specifically the differences between walking and running at the 1.8 m/s speed condition for all outcome variables. Additionally, bivariate correlation analysis was performed between joint stiffness (K_{joint}) and locomotion speeds in walking and running condition, as well as between joint stiffness (K_{joint}) and stance phase joint work (W_{joint}).

Chapter IV

To investigate age effects on lower extremity joint level kinematic and kinetic characteristics, and general gait patterns, calculation of different variables were conducted in both walking and running at different speeds. Joint kinematic variables include: joint ground contact angle (θ_{joint}^{GCA}) and toe-off angle (θ_{joint}^{TOA}) in stance, joint peak extension angle (θ_{joint}^{PEA}), joint peak flexion angle (θ_{joint}^{PFA}) and joint angle range of motion (θ_{joint}^{ROM}) over whole gait cycle. Joint kinetic variables include: joint stiffness (K_{joint}), stance and swing phase joint positive work (W_{joint}^{+}) and negative work (W_{joint}^{-}), stance phase joint extensor moment angular impulse (I_{joint}), total lower extremity support torque (I_{total}). General gait variables include: step length, step width, stance time, swing time and gait cadence. A more in-depth explanation of the variables selection and calculation details is given in Chapter IV. All the outcome variables were calculated and averaged from both limbs, normalized to body mass (where appropriate) and averaged across 3 gait cycles.

Joint stiffness (K_{joint}), joint work (W_{joint}), angular impulse (I_{joint}) and all joint kinematic variables were compared in a 2-way mixed effects ANOVAs (group × speed) for each joint, within walking and running conditions, respectively in SPSS. The factor of Group (young vs. middle age subjects) was tested for between subject effect and speed was tested for within subject effect in the statistical analysis. Initial alpha level was set to 0.05. When main effects or interaction effects were detected, Bonferroni adjustments were used for pairwise comparison. Follow up pairwise comparison alpha level was set to 0.05 divided by the number of comparisons. An unpaired sample t-test was conducted to test general gait pattern variables (step length, step width, stance time, swing time and gait cadence) in each speed between young and middle age group.

Chapter V

To investigate lower extremity joint kinetic patterns in gait transition process, all outcome variables calculation and analysis were focused on the two steps before gait transition (S-2, S-1), the transition step (S0) and the two steps after transition (S1, S2) for both WRT and RWT. Joint kinetic variables include: joint stiffness (K_{joint}), stance and swing phase joint positive work (W^{+}_{joint}) and negative work (W^{-}_{joint}), stance phase joint extensor moment angular impulse (I_{joint}), total lower extremity support torque (I_{total}), and joint stance and swing phase peak extension and flexion power. A more in-depth explanation of the variables selection and calculation details is given in Chapter V.

Joint stiffness (K_{joint}), joint work (W_{joint}) and angular impulse (I_{joint}) were examined for differences between joints and steps before, during and after the transition using a 2-way ANOVAs (joint × step) for WRT and RWT in SPSS, respectively. Total support torque (I_{total}), joint stance and swing phase peak extension and flexion power were examined using a 1-way ANOVA to compare between the five steps tested during WRT and RWT, respectively. For this analysis, peak joint extension and flexion power analysis was conducted within ankle, knee and hip, separately. Initial alpha level was set to 0.05. When main effects or interaction effects were detected, Bonferroni adjustments were used for pairwise comparison, so that the alpha level was divided by the number of comparisons.

Chapter VI

To investigate whole body COM dynamic patterns in running, a fifteen-segment whole-body model was built in Visual 3D. Whole body COM position was calculated as the weighted sum of 15 segments of the whole-body model. Outcome variables include: vertical stiffness (K_{vert}), leg-spring stiffness (K_{leg}), joint stiffness (K_{joint}), COM instantaneous power (P_{com}), COM positive and negative mechanical work (W_{com}). A more in-depth explanation of the variables selection and calculation details is given in Chapter VI. All outcome variables were calculated and averaged from both limbs and averaged across three gait cycles.

Vertical stiffness (K_{vert}), leg stiffness (K_{leg}), joint stiffness (K_{joint}), COM positive work (W_{com}^{+}) and negative work (W_{com}^{-}) were examined using a 1-way ANOVA to compare among six speeds in SPSS. Initial alpha level was set to 0.05. When main effect was detected, Bonferroni adjustments were used for pairwise comparison, so that the alpha level was divided by the number of comparisons (adjusted α = 0.0033 for all variables' pairwise comparison in this study). Additionally, multiple linear regression analysis was conducted to develop models for predicting K_{vert}, K_{leg} from K_{joint} (ankle, knee and hip joint stiffness) within each running speed, respectively in SPSS. Lastly, simple linear regression analysis was used to examine the relationship between sagittal plane COM positive work (W_{coms}^{+}) and K_{vert}, K_{leg} respectively across speeds, to investigate whether W_{coms}^{+} could be predicted from K_{vert} or K_{leg}.

CHAPTER III

MODULATION OF LOWER EXTREMITY JOINT STIFFNESS, WORK AND POWER AT DIFFERENT WALKING AND RUNNING SPEEDS

This work was published in volume 58 of the *Human Movement Science* in January 2018. Li Jin designed this study, collected the data and analyzed it. Michael E. Hahn provided mentorship activities, including assistance with study design, general oversight of the project, and editing and finalizing of the journal manuscript.

Introduction

Locomotion is an important function in human activities, and walking and running are the primary forms. Both activities require complex coordination between different muscles, tendons and ligaments (Ferris, Louie, & Farley, 1998; Kuo, 2007), associated with kinematic and kinetic pattern changes of joints and segments (Li Li, Van Den Bogert, Caldwell, Van Emmerik, & Hamill, 1999). Walking can be loosely described as 'vaulting' over relatively stiff legs (Geyer et al., 2006), while running is often described as a bouncing movement, on 'springy' legs (Geyer et al., 2006; McGowan et al., 2012). To better investigate and interpret the dynamics of walking and running, a simplified inverted pendulum model (Adamczyk & Kuo, 2009; Kuo, 2007; McGrath et al., 2015) and spring mass model (Alexander, 1992; C T Farley et al., 1993; Claire T Farley & Gonzalez, 1996; McGowan et al., 2012; McMahon & Cheng, 1990) have both been used in previous research. In human locomotion, the lower extremity can be regarded as a system requiring joint level dynamic pattern coordination. When locomotion speeds change, lower extremity joints serve different functional roles (Qiao & Jindrich, 2016). Faster locomotion speeds coincide with an increase in kinetic energy of the whole body, due to more joint level mechanical work and power being generated than is absorbed (Schache et al., 2015). When walking at continuously increasing speeds, a spontaneous walk to run transition occurs at a fairly predictable speed (around 2.17 m/s) (Veerle Segers et al., 2007). Gait transition speed is influenced by how speed changes are introduced and there appears to be a redistribution of joint level mechanical work among the lower extremity joints (Farris & Sawicki, 2012). Change of locomotion speed may require different strategies, within one locomotion state and between different locomotion states (e.g., gait transition). A more detailed investigation is needed to better understand how joint level functional roles and mechanical patterns are coordinated among the different phases of locomotion, in response to changes in speed. A deeper understanding of joint level mechanics and functional interactions will benefit rehabilitation programs and assistive device development.

Joint level mechanics during locomotion requires the elastic potential characteristics of the musculotendinous system to absorb energy during the braking phase of early stance and generate energy during the propulsive phase in late stance (G. A. Cavagna, 1977; Kuitunen et al., 2002). The stretching and shortening phenomenon under loading conditions in different joints during locomotion has been described as having spring-like behavior (Kuitunen et al., 2002). Intersegmental displacement as a function of joint moment has been defined as dynamic joint stiffness (Crenna & Frigo, 2011; Davis & DeLuca, 1996; Gabriel et al., 2008). Previous studies have compared lower extremity joint stiffness in walking and running. The ankle joint moment-angle relationship has been investigated between males and females, as well as across different age ranges (Crenna & Frigo, 2011; Gabriel et al., 2008). It was reported that male subjects tended to have a higher ankle joint stiffness and higher joint work in normal walking, and ankle joint stiffness was not significantly different between different ages (Crenna & Frigo, 2011; Gabriel et al., 2008). Frigo et al. (1996) investigated the effect of different walking speeds on ankle, knee and hip joint angle-moment relationships. They reported that the various calculated slopes during different phases of plotted angle-moment relationships of each joint indicate speed dependence. However, findings of joint stiffness in running condition have been mixed. Ankle joint stiffness has been reported to remain unchanged when running from 2.5 – 6.5 m/s, as well as from 70% – 100% maximum running speed (Arampatzis et al., 1999; Kuitunen et al., 2002). However, it has also been reported that ankle joint stiffness was higher in sprinting compared to slower speed running (Stefanyshyn & Nigg, 1998). However, knee joint stiffness has been reported to increase with increased running speed (Arampatzis et al., 1999; Kuitunen et al., 2002). Knee joint stiffness tended to be higher than ankle joint stiffness and the knee joint was observed to have a higher magnitude of extension compared to the ankle joint in running (Günther & Blickhan, 2002). It remains that little is known about the concurrent stiffness patterns of lower extremity joints across locomotion speeds and between locomotion states. This study provides further information regarding joint stiffness patterns while walking and running at various speeds.

Modulation of joint level mechanical work and power is known to contribute to dynamic movement in different locomotion speeds (Ebrahimi, Goldberg, & Stanhope, 2017). Farris & Sawicki (2012) found that the relative contribution of the ankle, knee and hip to total positive power did not change across walking and running speeds. In other findings, lower extremity joint work and average power did not proportionally increase from walking to sprinting (Schache et al., 2015). Instead, the contribution to the total average power tended to transfer between joints as speed changed (Schache et al., 2015). However, Anahid Ebrahimi et al. (2017) reported that stance phase relative ankle joint positive and negative work increased with walking speeds. With these apparent contradictions, the relationship between stance phase joint level mechanical loading and response, and the specific functional roles played by the joints in energy generation and absorption between stance phase and swing phase in different locomotion speeds remains unclear. More detailed comparisons are needed about the transfer mechanisms used during stance and swing phase joint work and average power in both walking and running.

Joint stiffness and joint level energy exchange mechanics are regarded as two major aspects of dynamic joint function (Crenna & Frigo, 2011). Previous studies have investigated joint stiffness, joint work and power patterns separately in different activities, with occasionally contradictory findings. However, more information is needed about the combination of joint stiffness, work and power in both walking and running, to more fully understand the relationship between stance phase joint dynamic loading response (specifically, in braking and propulsion phases), joint mechanical work and average power generation and absorption, when locomotion tasks and speeds change. Additionally, this study provides a separate analysis of stance and swing phase joint work and average power, providing a more detailed view of lower extremity joint function in different phases of walking and running across different speeds. An increased understanding of these relationships should provide a better framework for future assistive device development, which may be suitable for multiple tasks of human locomotion and better emulate the functional behavior of human limbs (Shamaei, Sawicki, & Dollar, 2013). The purpose of this study was to investigate lower extremity joint level stiffness, stance and swing phase joint work and average power in walking and running across a range of speeds. We hypothesized that: (1) lower extremity joints stiffness would increase when locomotion speeds increased, and (2) joint stiffness, joint work and power would be higher in running compared with walking.

Methods

Recruitment

Ten abled-bodied subjects participated in the study (23 $\pm$ 5.3 years, 170 $\pm$ 11.2 cm, 67 $\pm$ 14.2 kg). All subjects signed informed written consent approved by the university's institutional review board before participation. Subjects were excluded based on any of the following criteria: a history of neurologic deficits or other musculoskeletal disorders that would affect gait, a history of rheumatic diseases, or a history of unexpected falls in the previous six months.

Study Design and Experimental Protocol

After measuring height and body mass of each subjects, 55 retro-reflective markers were placed on the skin surface, adapted from a previously published whole body marker set (Sawers & Hahn, 2012). Subjects were first instructed to walk on a force-instrumented treadmill (Bertec, Inc., Columbus, OH) at seven increasing speeds, from 0.8 to 2.0 m/s (at 0.2 m/s intervals), for 90 seconds per stage. Then they were asked to run at six different speeds, from 1.8 to 3.8 m/s (at 0.4 m/s intervals), for 75 seconds per stage. Walking conditions were tested before running conditions, and there was a break between walking and running conditions. Subjects were allowed to rest at any time during the testing.

Data Collection

Data were extracted from the middle strides (20 strides on average) of each stage. Segmental kinematic data were collected at 120 Hz using an 8-camera motion capture system (Motion Analysis Corp., Santa Rosa, CA). Ground reaction force data were collected at 1200 Hz using the force-instrumented treadmill. Kinematic and kinetic data were filtered with a low-pass fourth-order Butterworth filter at 6 Hz and 50 Hz, respectively.

Data Analysis

Lower extremity joint angles, moments and net powers were calculated using an inverse dynamics model coded in Visual 3D (C-Motion, Inc., Germantown, MD). Joint stiffness (K_{joint}) was calculated as a change in joint moment (ΔM_{joint}) divided by joint angular displacement ($\Delta\theta_{joint}$) in the braking phase of ground contact, based on the anterior-posterior ground reaction force to investigate stance phase joint loading response (Hobara et al., 2013; Kuitunen et al., 2002), expressed as: $K_{joint} = \Delta M_{joint}/\Delta\theta_{joint}$. Stance and swing phase joint positive work (W_{joint}^{+}) and negative work (W_{joint}^{-}) were calculated as the sum of all positive or negative net joint power integrated over time, respectively (Schache et al., 2015). Stance and swing phase joint average positive power (P^{+}_{joint}) and negative power (P^{-}_{joint}) were calculated as joint work (W_{joint}) divided by stance and swing phase time, respectively (Farris & Sawicki, 2012; Schache et al., 2015). All dependent variables (joint stiffness, stance phase and swing phase joint positive and negative work, stance phase and swing phase joint average positive and negative power) were calculated and averaged from both limbs, normalized to body mass and averaged across 3 gait cycles. Lastly, stance phase sagittal plane ankle angle and moment values were plotted using a custom written Matlab program (R2016b, Mathworks, Natick, MA), to examine the dynamic loading response and energy generation patterns in different locomotion tasks and speeds.

Statistical Analysis

Statistical analysis was performed using two 2-way ANOVAs (joint × speed); one for walking and one for running conditions, in SPSS (V22.0, IBM, Armonk, NY). Bonferroni adjustments were used for pairwise comparisons. Initial alpha level was set to 0.05. When main effect or interaction effect were detected, post hoc analyses were conducted. Follow up pairwise comparison alpha level was set to 0.05 divided by the number of comparisons. A paired sample t-test was conducted to test specifically the differences between walking and running at the 1.8 m/s speed condition for all outcome variables. Additionally, bivariate correlation analysis was performed between joint stiffness (K_{joint}) and locomotion speeds in walking and running condition, as well as between joint stiffness (K_{joint}) and stance phase joint work (W_{joint}).

Results

For K_{joint} in the walking speed conditions, there was a significant joint × speed interaction effect (p = .002), which led to a post-hoc pairwise comparison alpha level that was adjusted to 0.0006. K_{knee} was significantly greater than K_{hip} at 0.8 m/s (p < .0005, Fig. 3.1). In running speed conditions, stiffness values of all three joints were positively associated with running speed (p < .05, r = 0.96 for ankle, r = 0.98 for knee, r = 0.82 for hip) (Fig. 3.2). Additionally**,** joint main effect (p = .0002) and speed main effect (p < .0001) were both significant, interaction effect was not significant (p = .81), and joint level pairwise comparison was conducted (adjusted α = 0.0167). K_{knee} was lower than K_{ankle} (p = .004) and K_{hip} (p = .005) across all running speeds (Fig. 3.2). The K_{joint} at 1.8 m/s speed was compared between walking and running conditions via paired t-test. We found that K_{ankle} (p = .002) and K_{hip} (p < .001) were higher in running compared with walking.

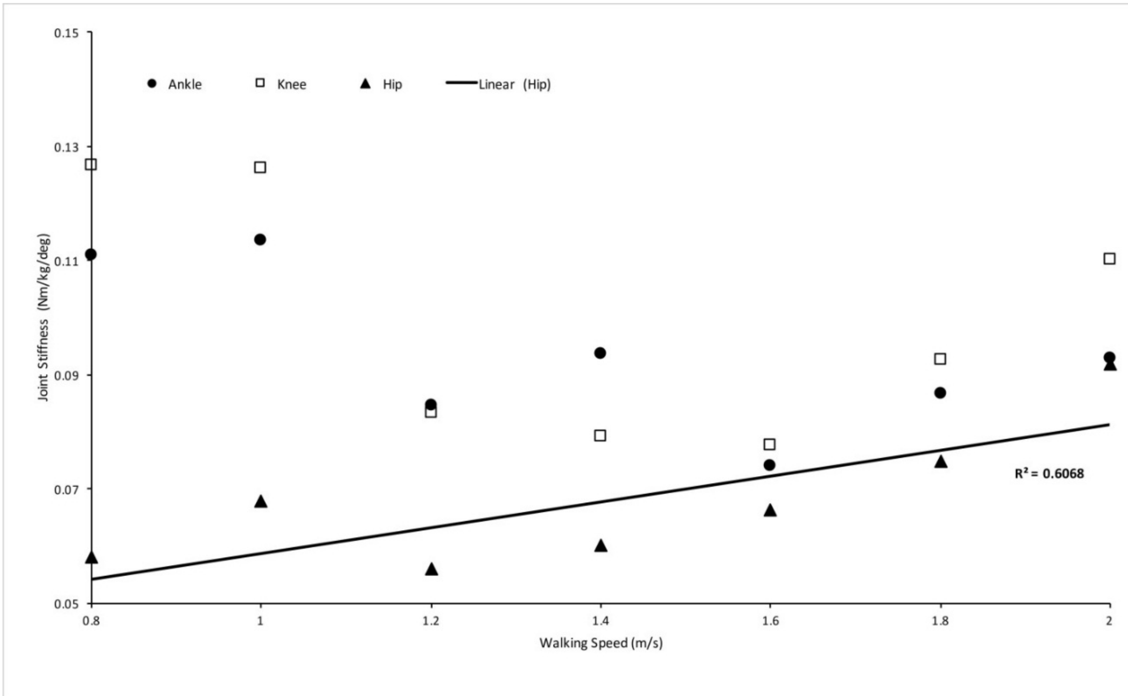

Fig 3.1. Joint stiffness in walking condition.

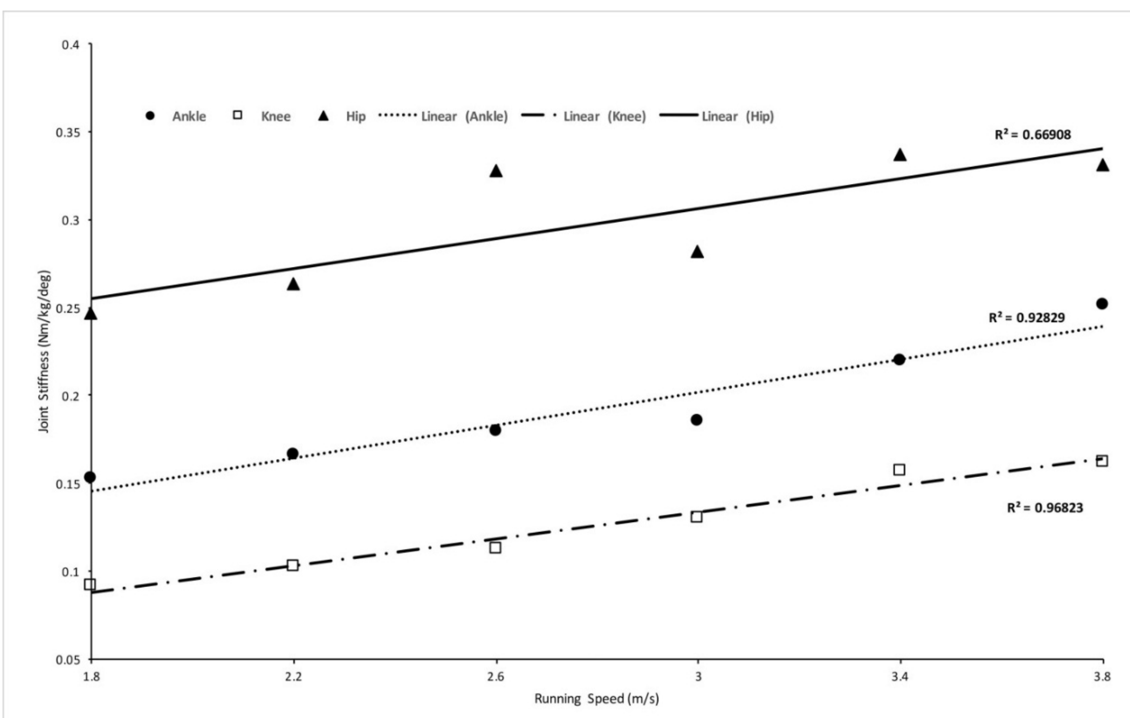

Fig 3.2. Joint stiffness in running condition.

During walking stance phase, the positive work and positive power of all three joints had a similar positive association with walking speeds (Table 3.1, Table 3.3). The ankle joint performed more positive work and generated more power than the knee and hip joint across walking speeds. For joint positive work, the joint × speed interaction effect was significant (p = .045), and pairwise comparison was conducted (adjusted α = 0.0006). Additionally, joint positive power had a significant joint × speed interaction effect (p = .008), and pairwise comparison was conducted (adjusted α = 0.0006). At 1.4 m/s, W^{+}_{ankle} was higher than W^{+}_{knee} (p = .0003) and W^{+}_{hip} (p < .0001) (Table 3.1), and P^{+}_{ankle} was higher than P^{+}_{knee} (p = .0003) and P^{+}_{hip} (p < .0001) (Table 3.3). Further, W^{+}_{ankle} at 1.4 m/s was higher than at 0.8 m/s (p = .0002), while W^{+}_{hip} at 2.0 m/s was higher than at 0.8 m/s (p = .0002) (Table 3.1). When walking speed increased, W^{-}_{ankle} tended to decrease while W^{-}_{knee} and W^{-}_{hip} tended to increase (Table 3.1). With increased walking speed P^{-}_{ankle} remained unchanged, and P^{-}_{knee} and P^{-}_{hip} tended to increase (Table 3.3).

Table 3.1. Joint work across different walking speeds. Sample Mean (SD); n = 10.

Joint Work (J/kg)	**Walking Speeds (m/s)**						
	0.8	**1.0**	**1.2**	**1.4**	**1.6**	**1.8**	**2.0**
Stance Phase Positive Work							
Ankle	0.13 (0.06)[c]	0.18 (0.08)	0.18 (0.08)	0.25 (0.07)[a,b,c]	0.27 (0.12)	0.31 (0.12)	0.34 (0.16)
Knee	0.10 (0.07)	0.10 (0.07)	0.11 (0.04)	0.13 (0.03)[a]	0.18 (0.07)	0.22 (0.10)	0.22 (0.14)
Hip	0.09 (0.05)[d]	0.13 (0.05)	0.09 (0.04)	0.12 (0.03)[b]	0.13 (0.05)	0.16 (0.07)	0.18 (0.08)[d]
Stance Phase Negative Work							
Ankle	0.24 (0.07)	0.22 (0.05)	0.16 (0.09)	0.18 (0.05)	0.13 (0.05)	0.13 (0.05)	0.11 (0.03)
Knee	0.07 (0.06)	0.14 (0.14)	0.21 (0.18)	0.15 (0.03)	0.27 (0.24)	0.33 (0.26)	0.24 (0.08)
Hip	0.07 (0.06)	0.11 (0.13)	0.23 (0.20)	0.13 (0.07)	0.32 (0.35)	0.36 (0.41)	0.22 (0.06)
Swing Phase Positive Work							
Ankle[e]	<0.01 (0.00)	<0.01 (0.00)	0.01 (0.00)	0.01 (0.00)	0.01 (0.00)	0.01 (0.00)	0.01 (0.00)
Knee	0.01 (0.01)	<0.01 (0.00)	<0.01 (0.00)	0.01 (0.00)	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)
Hip[e]	0.04 (0.02)	0.06 (0.03)	0.08 (0.03)	0.08 (0.01)	0.10 (0.03)	0.13 (0.03)	0.16 (0.04)
Swing Phase Negative Work							
Ankle[f]	<0.01 (0.00)	<0.01 (0.00)	<0.01 (0.00)	<0.01 (0.00)	<0.01 (0.00)	<0.01 (0.00)	<0.01 (0.00)
Knee[f,g]	0.10 (0.02)	0.13 (0.02)	0.16 (0.02)	0.17 (0.01)	0.18 (0.04)	0.22 (0.03)	0.26 (0.03)
Hip[g]	<0.01 (0.01)	<0.01 (0.00)	<0.01 (0.00)	<0.01 (0.00)	0.01 (0.01)	0.01 (0.01)	0.02 (0.02)

Note: <0.01 indicates a negligible value; Joint negative work data were presented in absolute values.
a: Statistically significant differences of stance phase positive work between ankle and knee joint at 1.4 m/s, (p = .0003);
b: Differences of stance phase positive work between ankle and hip joint at 1.4 m/s, ($p < .0001$);

c: Differences of stance phase ankle joint positive work between 0.8 m/s and 1.4 m/s, (p = .0002);
d: Differences of stance phase hip joint positive work between 0.8 m/s and 2.0 m/s, (p = .0002);
e: Differences of swing phase positive work between ankle and hip joint in each walking speed, ($p < .0006$);
f: Differences of swing phase negative work between ankle and knee joint across all walking speeds, ($p < .0001$);
g: Differences of swing phase negative work between hip and knee joint across all walking speeds, ($p < .0001$).

Table 3.2. Joint work across different running speeds. Sample Mean (SD); n = 10.

Joint Work (J/kg)	**Running Speeds (m/s)**					
	1.8	**2.2**	**2.6**	**3.0**	**3.4**	**3.8**
Stance Phase Positive Work						
Ankle[a,b]	0.46 (0.18)	0.51 (0.21)	0.46 (0.26)	0.50 (0.24)	0.52 (0.28)	0.70 (0.17)
Knee[a,c]	0.20 (0.06)	0.23 (0.05)	0.24 (0.06)	0.30 (0.11)	0.29 (0.13)	0.30 (0.10)
Hip[b,c]	0.04 (0.02)	0.05 (0.03)	0.13 (0.16)	0.14 (0.08)	0.21 (0.13)	0.21 (0.05)
Stance Phase Negative Work						
Ankle	0.30 (0.09)	0.31 (0.10)	0.31 (0.09)	0.35 (0.14)	0.33 (0.14)	0.44 (0.07)
Knee	0.37 (0.17)	0.38 (0.13)	0.42 (0.22)	0.46 (0.19)	0.45 (0.14)	0.36 (0.05)
Hip	0.11 (0.07)	0.12 (0.07)	0.13 (0.11)	0.18 (0.11)	0.20 (0.16)	0.22 (0.17)
Swing Phase Positive Work						
Ankle[d]	0.01 (0.00)	0.01 (0.00)	0.01 (0.00)	0.01 (0.00)	0.01 (0.00)	0.02 (0.00)
Knee[e]	0.01 (0.01)	0.01 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.02)	0.03 (0.03)
Hip[d,e]	0.15 (0.02)	0.22 (0.05)	0.32 (0.06)	0.44 (0.11)	0.56 (0.12)	0.67 (0.14)
Swing Phase Negative Work						
Ankle[f,h]	<0.01 (0.00)	<0.01 (0.00)	<0.01 (0.00)	<0.01 (0.00)	0.01 (0.01)	0.01 (0.00)
Knee[f,g]	0.26 (0.03)	0.35 (0.03)	0.47 (0.06)	0.58 (0.07)	0.73 (0.08)	0.88 (0.14)
Hip[g,h]	0.01 (0.01)	0.02 (0.02)	0.04 (0.02)	0.05 (0.02)	0.08 (0.03)	0.08 (0.04)

Note: <0.01 indicates a negligible value; Joint negative work data were presented in absolute values.
a: Statistically significant differences of stance phase joint positive work between ankle and knee joint

across all running speeds, ($p = .003$);
b: Differences of stance phase positive work between ankle and hip joint across all running speeds, ($p = .0001$);
c: Differences of stance phase positive work between knee and hip joint across all running speeds, ($p = .008$);
d: Differences of swing phase positive work between ankle and hip joint across all running speeds, ($p < .0001$);
e: Differences of swing phase positive work between knee and hip joint across all running speeds, ($p < .0001$);
f: Differences of swing phase negative work between ankle and knee joint in each running speed, ($p < .0001$);
g: Differences of swing phase negative work between hip and knee joint in each running speed, ($p < .0001$);
h: Differences of swing phase negative work between ankle and hip joint when running between 2.2 – 3.4 m/s, ($p < .0001$).

Table 3.3. Joint average power across different walking speeds. Sample Mean (SD); n = 10.

Joint Power (W/kg)	**Walking Speeds (m/s)**						
	0.8	**1.0**	**1.2**	**1.4**	**1.6**	**1.8**	**2.0**
Stance Phase Positive Power							
Ankle	0.17 (0.07)	0.25 (0.10)	0.28 (0.11)	0.40 (0.11)[a,b]	0.46 (0.20)	0.59 (0.25)	0.67 (0.30)
Knee	0.13 (0.10)	0.15 (0.10)	0.18 (0.07)	0.21 (0.04)[a]	0.31 (0.13)	0.43 (0.24)	0.43 (0.24)
Hip	0.12 (0.07)	0.19 (0.07)	0.14 (0.06)	0.19 (0.05)[b]	0.23 (0.10)	0.30 (0.13)	0.37 (0.18)
Stance Phase Negative Power							
Ankle	0.31 (0.10)	0.31 (0.06)	0.25 (0.14)	0.29 (0.09)	0.22 (0.09)	0.25 (0.09)	0.21 (0.07)
Knee	0.09 (0.07)	0.20 (0.20)	0.33 (0.31)	0.24 (0.06)	0.47 (0.44)	0.63 (0.55)	0.49 (0.15)
Hip	0.09 (0.07)	0.16 (0.18)	0.37 (0.35)	0.21 (0.11)	0.57 (0.64)	0.71 (0.90)	0.45 (0.13)
Swing Phase Positive Power							
Ankle[c]	0.01 (0.01)	0.01 (0.00)	0.01 (0.00)	0.01 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)
Knee[d]	0.02 (0.03)	0.01 (0.01)	0.01 (0.00)	0.01 (0.00)	0.02 (0.01)	0.02 (0.01)	0.02 (0.02)
Hip[c,d]	0.08 (0.04)	0.12 (0.09)	0.16 (0.07)	0.17 (0.06)	0.22 (0.09)	0.27 (0.07)	0.37 (0.15)
Swing Phase Negative Power							
Ankle	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.02)	0.02 (0.05)	0.02 (0.04)	0.01 (0.03)
Knee	0.18 (0.05)	0.25 (0.06)	0.31 (0.09)	0.35 (0.10)	0.38 (0.12)	0.48 (0.14)	0.60 (0.14)
Hip	0.01 (0.01)	<0.01 (0.00)	<0.01 (0.00)	0.01 (0.00)	0.02 (0.01)	0.03 (0.02)	0.06 (0.03)

Note: <0.01 indicates a negligible value; Joint negative power data were presented in absolute values.

a: Statistically significant differences of stance phase positive power between ankle and knee joint at 1.4 m/s, (p = .0003);
b: Differences of stance phase positive power between ankle and hip joint at 1.4 m/s, (p < .0001);
c: Differences of swing phase positive power between ankle and hip joint in each speed (except walking at 1.0 m/s), (p <

.0006);
d: Differences of swing phase positive power between knee and hip joint in each speed (except walking at 1.0 m/s), ($p <$ .0006).

In the stance phase of running, positive work joint main effect ($p < .0001$) and speed main effect ($p < .0001$) were both significant, the interaction effect was not significant ($p = .66$), and a joint level pairwise comparison was conducted (adjusted α = 0.0167). Positive power joint main effect ($p < .001$) and speed main effect ($p < .001$) were both significant, however the interaction effect was not significant ($p = .22$), and a joint level pairwise comparison was conducted (adjusted α = 0.0167). W_{ankle}^{+} was higher than W_{knee}^{+} ($p = .003$) and W_{hip}^{+} ($p = .0001$) across all speeds, and W_{knee}^{+} was also higher than W_{hip}^{+} ($p = .008$) (Table 3.2). Similarly, P_{ankle}^{+} was higher than P_{knee}^{+} ($p = .004$) and P_{hip}^{+} ($p = .0003$) across all speeds (Table 3.4). When running speeds increased, negative work and power of all three joints tended to increase and the hip joint tended to be smaller than ankle and knee joint.

Table 3.4. Joint average power across different running speeds. Sample Mean (SD); n = 10.

Joint Power (W/kg)	**Running Speeds (m/s)**					
	1.8	**2.2**	**2.6**	**3.0**	**3.4**	**3.8**
Stance Phase Positive Power						
Ankle[a,b]	1.47 (0.58)	1.86 (0.65)	1.87 (0.96)	2.15 (1.01)	2.45 (1.22)	3.54 (0.77)
Knee[a]	0.65 (0.22)	0.88 (0.24)	1.00 (0.24)	1.31 (0.49)	1.39 (0.67)	1.54 (0.56)
Hip[b]	0.13 (0.05)	0.19 (0.09)	0.62 (0.86)	0.60 (0.35)	1.02 (0.74)	1.06 (0.26)
Stance Phase Negative Power						
Ankle	1.04 (0.36)	1.19 (0.32)	1.38 (0.44)	1.57 (0.58)	1.59 (0.59)	2.21 (0.37)
Knee	1.23 (0.69)	1.41 (0.41)	1.84 (1.12)	2.08 (1.04)	2.21 (0.86)	1.81 (0.34)
Hip	0.37 (0.30)	0.48 (0.31)	0.59 (0.45)	0.78 (0.46)	0.99 (0.74)	1.12 (0.92)
Swing Phase Positive Power						
Ankle	0.02 (0.00)	0.02 (0.01)	0.02 (0.00)	0.02 (0.01)	0.02 (0.00)	0.03 (0.01)
Knee	0.01 (0.01)	0.02 (0.02)	0.03 (0.02)	0.03 (0.02)	0.03 (0.03)	0.04 (0.04)
Hip	0.28 (0.05)	0.38 (0.10)	0.52 (0.14)	0.72 (0.24)	0.93 (0.25)	1.04 (0.24)
Swing Phase Negative Power						
Ankle	<0.01 (0.00)	<0.01 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.00)
Knee	0.47 (0.03)	0.59 (0.15)	0.78 (0.17)	0.95 (0.25)	1.20 (0.23)	1.35 (0.26)
Hip	0.02 (0.01)	0.04 (0.04)	0.06 (0.04)	0.08 (0.03)	0.13 (0.07)	0.13 (0.06)

Note: <0.01 indicates a negligible value; Joint negative power data were presented in absolute values.
a: Statistically significant differences of stance phase positive power between ankle and knee joint across all running speeds, (p = .004);

b: Differences of stance phase positive power between ankle and hip joint across all running speeds, (p = .0003).

During the swing phase of walking, positive work, negative work and positive power joint × speed interaction effect were significant ($p < .0001$), respectively, and pairwise comparison was conducted (adjusted α = 0.0006). W_{hip}^{+} was higher than W_{ankle}^{+} in each walking speed ($p < .0006$), and W_{hip}^{+} tended to increase when walking speeds increased (Table 3.1). Except at 1.0 m/s, P_{hip}^{+} was higher than P_{ankle}^{+} and P_{knee}^{+} in all other speeds ($p < .0006$) (Table 3.3). W_{knee}^{-} was higher than W_{ankle}^{-} and W_{hip}^{-} across all speeds ($p < .0001$), and W_{knee}^{-} had a positive association with walking speeds (Table 3.1). A similar trend was found in P_{knee}^{-} in walking as well (Table 3.3).

In the swing phase of running, positive work, negative work and positive power joint × speed interaction effect were significant ($p < .0001$), respectively, and pairwise comparison was conducted (adjusted α = 0.0008). W_{hip}^{+} was higher than W_{ankle}^{+} and W_{knee}^{+} across all speeds ($p < .0001$), and W_{hip}^{+} tended to have a positive association with running speeds (Table 3.2). A similar trend was observed for P_{hip}^{+} across running speeds (Table 3.4). For joint negative work, W_{knee}^{-} was higher than W_{ankle}^{-} and W_{hip}^{-} in each running speed ($p < .0001$). Between 2.2 – 3.4 m/s, W_{hip}^{-} was higher than W_{ankle}^{-} as well ($p < .0001$) (Table 3.2). When running speeds increased, W_{knee}^{-} tended to increase (Table 3.2). P_{knee}^{-} also had a similar pattern in the swing phase of running (Table 3.4).

We compared W_{joint} and P_{joint} between walking and running at 1.8 m/s via paired t-test. In stance phase, W_{ankle}^{+} ($p = .04$) and P_{ankle}^{+} ($p < .001$) were higher in running compared with walking, while W_{hip}^{+} ($p < .001$) and P_{hip}^{+} ($p = .006$) were higher in the walking condition. The W_{ankle}^{-} ($p = .001$) and P_{ankle}^{-} ($p < .001$) were higher during the stance phase running as well. In swing phase, W_{knee}^{-} (p = .001) was higher in running compared to walking at 1.8 m/s.

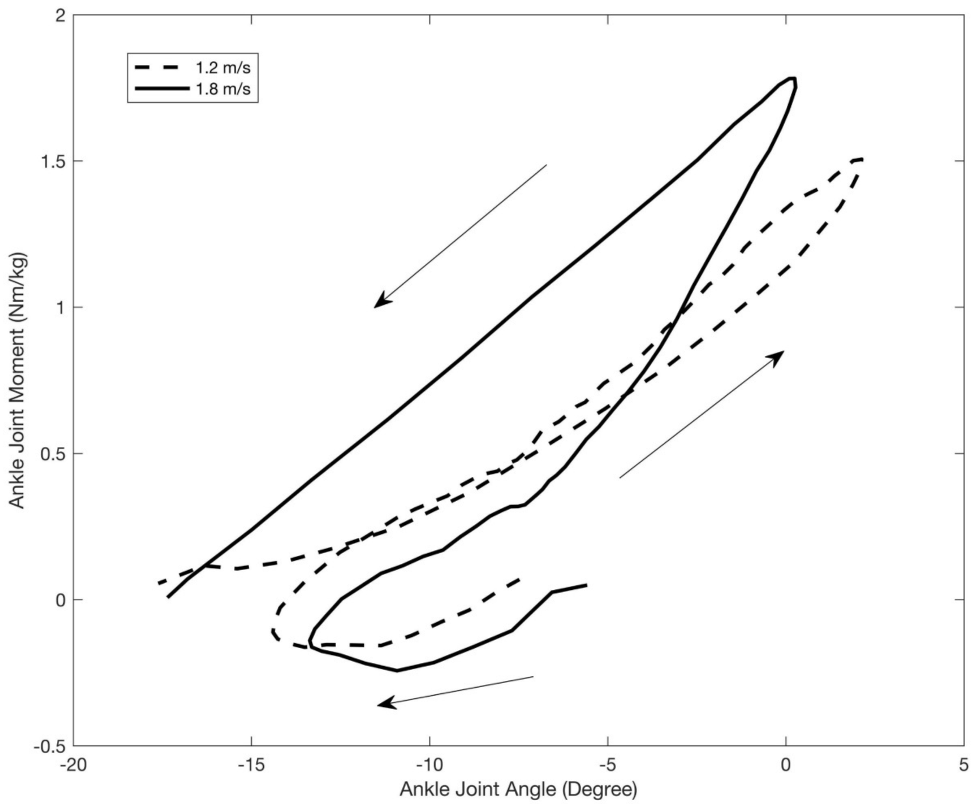

Fig 3.3. Stance phase ankle angle-moment curve for two different walking speed conditions from one exemplar subject (averaged over 3 trials).

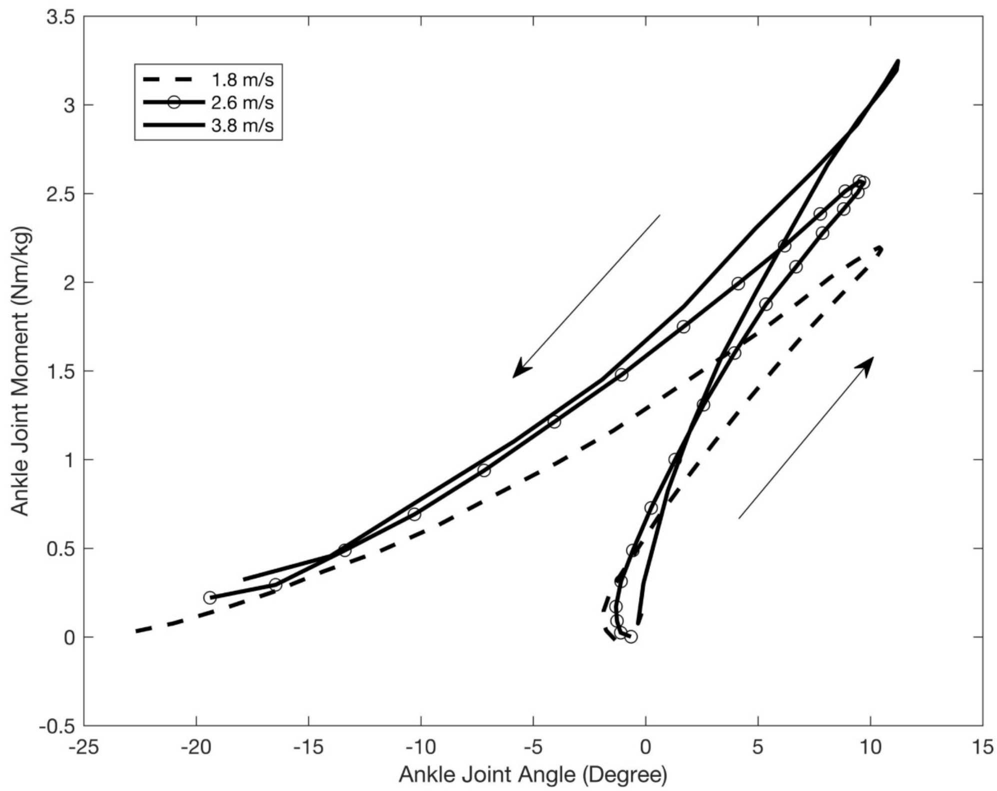

Fig 3.4. Stance phase ankle angle-moment curve for three different running speed conditions from one exemplar subject (averaged over 3 trials).

Discussion

The purpose of this study was to investigate lower extremity joint level stiffness, stance and swing phase joint work and average power in walking and running across a range of speeds. The initial hypothesis that locomotion tasks and speed changes would have influence on joint level mechanics patterns was partially supported. Specifically, K_{ankle} and K_{hip} were higher in the running condition compared with walking at the same locomotion speed (1.8 m/s) (Fig. 3.1, Fig. 3.2). During slow walking speed conditions (0.8 – 1.0 m/s), step length was small, ankle and knee angular displacement was relatively small. These findings likely contributed to the result of K_{ankle} and K_{knee}

remaining relatively high during the braking phase. In the more typical walking speed range (1.2 – 1.6 m/s), knee joint angular displacement increased, with an associated decrease in K_{knee}. However, in the faster walking speed range (1.6 – 2.0 m/s), increased sagittal plane joint moment coupled with decreased angular displacement, resulted in greater sagittal plane joint moments and increased K_{joint}.

In the running conditions, K_{knee} increased with speed. This finding was in agreement with previous studies (Arampatzis et al., 1999; Kuitunen et al., 2002). Interestingly, K_{knee} was lowest among the three joints, however W_{knee}^{-} was greater during the stance phase of running. This indicates that the knee joint performs more energy absorption and plays a coordination role between the ankle and hip joints during stance. In the early stance phase (also known as the braking phase), the effect of mechanical loading from the distal force through the ankle joint, concurrent with body weight loading through the hip joint has a compounded effect on the knee joint. Knee flexion can store elastic energy in early stance phase to be returned in late stance phase with a stretch-reflex response leading to knee extension and energy return during push-off (Kuitunen et al., 2002).

We also investigated the ankle joint angle-moment relationship in the stance phase of walking and running. Similar to previous studies (Crenna & Frigo, 2011; Gabriel et al., 2008; Kuitunen et al., 2002), we observed a counterclockwise hysteresis loop (Fig. 3.3, Fig. 3.4). For the walking conditions, there were 3 apparent phases: initial contact, ascending phase and descending phase (Crenna & Frigo, 2011). Within the initial contact and ascending phase, the area under the curve can be regarded as the negative mechanical work performed by the ankle joint, based on the integral of moment over range of rotation principle (Crenna & Frigo, 2011). For the descending phase, the area under the curve is regarded as the positive mechanical work performed by the joint (Crenna & Frigo, 2011). When comparing the ankle angle-moment relationship between normal (1.2 m/s) and fast walking speed (1.8 m/s), the shape of the loop transformed from a narrow loop to the relatively large one. This change indicates that the ankle joint is generating more mechanical energy than is absorbed (Gabriel et al., 2008). In the present study, this observed change tended to have a larger effect when walking speeds increased (Fig. 3.3, Table 3.1). When translated to future assistive device development, this observation would suggest that an enhanced energy generation system would be needed as locomotion speeds increased, to better emulate the natural capacity of the biological ankle joint system.

The ankle joint angle-moment relationship was a little different in the running condition, where there were only two phases: an ascending and a descending phase (Fig. 3.4). When running speeds increased, the slope of the ascending phase and descending phase curve increased (Fig. 3.4), indicating there was more of both mechanical energy absorption and generation. When comparing the ankle joint angle-moment relationship between walking and running at 1.8 m/s, there was a higher angular displacement in the ascending phase during walking (Fig. 3.3, Fig. 3.4). This resulted in a higher K_{ankle} value in running compared with walking at the same speed. For the descending phase, there tended to be a larger area under the curve in the running conditions, indicating that running requires a higher amount of stance phase energy generation compared with walking at the same speed. For future ankle assistive device or ankle prosthesis development, it will be necessary to consider that running conditions require greater ankle joint stiffness and a higher amount of energy generation compared with walking at the same speed. Moreover, the ankle angle-moment relationship can provide valuable information about stance phase ankle joint loading and unloading patterns, and the connection between joint stiffness and joint level energy generation and absorption. Additionally, in the running conditions, stance phase W_{ankle}^{+} had strong positive association with K_{ankle} ($p < .0001$). This suggests that when running at slow to moderate speeds, higher ankle joint stiffness coincides with the musculotendinous stretch-reflex response of the joint to generate more energy in stance phase.

In the stance phase of walking and running, the ankle joint produced more positive work and power compared with the knee and hip joints across different speeds (Table 3.1, Table 3.2). This observation was in agreement with a previous study (Schache et al., 2015). It is clear that the ankle joint is important to generate energy in stance phase. When walking speeds increased, the knee and hip joint tended to absorb more energy, while the ankle joint absorbed less energy (Table 3.1). This suggests that when walking speeds increase, stance phase energy absorption among lower extremity joints tends to transfer from the distal segments to the proximal. In the stance phase of running, the ankle and knee tended to absorb more energy than the hip joint across speeds (Table 3.2). When locomotion speed increased, the positive work in all three joints increased in stance phase, similar to previous findings (Anahid Ebrahimi et al., 2017; Schache et al., 2015). In the swing phase of walking and running, the hip joint played a dominant role in generating positive work and power, while the knee joint played an energy absorption role.

Limitations

One limitation of the study is common to treadmill walking and running, as the locomotion speeds were controlled, and different subjects may have individual variations which may have been masked by the controlled speeds. The other limitation is that we assumed gait symmetry between left and right leg in walking and running condition. In cases where there is pronounced asymmetry, the individual distribution of joint work and power may vary beyond what was observed in the present study.

Future Work

For future studies, it would be beneficial to further investigate the relationship between joint stiffness and whole leg stiffness during different locomotion tasks and speeds. Additionally, it will be interesting to further examine the relationship between whole body center of mass dynamics and lower extremity joint mechanics during different locomotion speeds and transitions between speeds.

Conclusion

Lower extremity joint stiffness values were different between walking and running across speeds. Locomotion speed is an important factor which affects characteristics of joint level mechanical performance, in both stance and swing phase. When speeds increased in walking and running, stance phase ankle joint positive work, swing phase knee negative work and hip positive work tended to increase. Ankle joint was determined to be critical during stance phase for energy generation in walking and running over a range of speeds. Higher ankle joint stiffness results in more positive work performed and power generation in running. From the findings of this study it is apparent that different locomotion tasks (walking vs. running) can fundamentally change joint level mechanics, even at the same locomotion speed.

Bridge

The results of this study indicate change of speeds could significantly affect joint level kinetic characteristics within both walking and running locomotion states. Whether progression of age would also affect lower extremity joint level kinematic and kinetic patterns in both walking and running activities needs further investigation. The data set in this chapter is also included in Chapter IV for further age comparison analysis (young vs. middle age). It should be noted that we defined ankle joint sagittal plane dorsiflexion movement as positive in the ankle joint angle-moment curve in this Chapter, compared with joint neutral position. In Chapters IV and V we defined ankle joint sagittal plane dorsiflexion direction as negative, compared with joint neutral position.

CHAPTER IV

COMPARISON OF LOWER EXTREMITY JOINT MECHANICS AND GAIT PATTERNS BETWEEN YOUNG AND MIDDLE AGE PEOPLE IN WALKING AND RUNNING GAIT

This Chapter is currently unpublished. Li Jin designed this study, collected the data and analyzed it. Michael E. Hahn provided mentorship activities, including assistance with study design, general oversight of the project, and editing and finalizing of the journal manuscript.

Introduction

Human locomotion requires integration of physiological and biomechanical factors (Chung & Wang, 2010). These factors are affected by increasing age, and thus can influence changes in gait patterns. A decrease in preferred locomotion speed and step length have been reported as typical gait pattern changes associated with age increase (Judge et al., 1996; Kerrigan et al., 1998; Silder et al., 2008; Winter et al., 1990). This may indicate that to achieve the same locomotion speed compared with young age people, there may be some compensatory mechanisms among lower extremity joints and segments in middle-age and elderly individuals. These compensatory mechanisms are likely associated with relevant gait kinematic and kinetic pattern changes.

Previous studies have investigated the age effect on joint level kinematic and kinetic patterns in both walking (Anderson & Madigan, 2014; Boyer, Andriacchi, & Beaupre, 2012; Browne & Franz, 2017; Chung & Wang, 2010; DeVita et al., 2000;

Judge, Davis, & Ounpuu, 2017; Riley, Dellacroce, & Kerrigan, 2001; Silder et al., 2008) and running (Bus, 2003; Fukuchi & Duarte, 2008) gait. Older adults redistributed lower extremity joint moment and power to maintain similar gait performance compared with young people in walking (DeVita et al., 2000). Specifically, older adults tended to transfer the mechanical demands from the distal end of the lower extremity to the more proximal end, by increasing net positive work at the hip to compensate the decreased work generated at the ankle (Browne & Franz, 2017; DeVita et al., 2000; Silder et al., 2008). In running conditions, elderly people tended to have less knee joint range of motion and a loss of shock-absorbing capacity compared with young people (Bus, 2003; Fukuchi & Duarte, 2008). These observations indicate that the knee join may become a "stiffer" system as age increases. While most of these studies focused on comparisons between young and older adults, little is known about whether there are differences in gait patterns and lower extremity joint mechanics across a smaller age range: between young and middle-age people in walking and running conditions. Moreover, previous comparisons were mainly focused on self-selected walking (Boyer et al., 2012; Silder et al., 2008) and running (Bus, 2003) speeds. It remains unknown if there is an age effect on gait mechanics characteristics at the same locomotion task and speed condition. Specifically, in a range of control speed conditions in both walking and running, this study sought to determine if there are differences in gait kinematic and kinetic characteristics between young and middle age adults.

Joint torsional stiffness is a combination of joint level kinematic and kinetic variables in stance phase during locomotion. It reflects the sagittal plane dynamic loading response characteristics of a joint. Joint level stiffness has been reported to increase with age. This increase would result in relevant joint level kinetic changes, such as joint flexor/extensor moment and relevant stance phase extensor moment angular impulse, joint mechanical work, power generation and absorption (Silder et al., 2008). The purpose of this study was to investigate lower extremity joint level stiffness patterns, basic gait patterns and joint level kinematic and kinetic characteristics between young and middle age group in both walking and running, across speeds. Further, we sought to identify whether there is a compensatory mechanism among lower extremity joints in middle age people in a wide range of walking and running speeds. We hypothesized that the middle age group would have: (1) higher joint stiffness; (2) higher stance phase hip joint extensor moment angular impulse and positive work, lower ankle joint plantar flexor moment angular impulse and positive work; and (3) smaller joint angle range of motion, step length and higher gait cadence compared with young age group.

Methods

Recruitment

Ten young healthy subjects (23 $\pm$ 5.3 years, 170 $\pm$ 11.2 cm, 67 $\pm$ 14.2 kg) and ten middle age healthy subjects (51 $\pm$ 6.0 years, 173 $\pm$ 11.4 cm, 70 $\pm$ 15.0 kg) participated in the study. All subjects signed informed written consent approved by the university's institutional review board before participation. All subjects were without lower extremity musculoskeletal related injuries for the past 6 months before the test.

Study Design and Experimental Protocol

Fifty-five retro-reflective markers were placed on the skin surface of the subjects, based on a previously published whole body marker set (Sawers & Hahn, 2012). Subjects were first instructed to walk on a force-instrumented treadmill (Bertec, Inc., Columbus, OH) at seven increasing speeds, from 0.8 to 2.0 m/s (at 0.2 m/s intervals), for 90 seconds per stage. Subjects were then asked to run at six different speeds, from 1.8 to 3.8 m/s (at 0.4 m/s intervals), for 75 seconds per stage. Walking conditions were tested before running conditions, and subjects were given a break between walking and running conditions.

Data Collection

Data were extracted from the middle strides (20 strides on average) of each stage. Segmental kinematic data were collected at 120 Hz using an 8-camera motion capture system (Motion Analysis Corp., Santa Rosa, CA). Ground reaction force data were collected at 1200 Hz using the force-instrumented treadmill. Kinematic and kinetic data were filtered with a low-pass fourth-order Butterworth filter at 6 Hz and 50 Hz, respectively.

Data Analysis

Lower extremity joint angles, moments and net joint powers were calculated using an inverse dynamics model coded in Visual 3D (C-Motion, Inc., Germantown, MD). Joint stiffness (K_{joint}) was calculated as a change in sagittal plane joint moment (ΔM_{joint}) divided by sagittal plane joint angular displacement ($\Delta\theta_{joint}$) in the braking phase of ground contact, based on the anterior-posterior ground reaction force value

(Hobara et al., 2013; Kuitunen et al., 2002), expressed as: K_{joint} = $\Delta M_{joint}/\Delta\theta_{joint}$. Stance and swing phase joint positive work (W_{joint}^{+}) and negative work (W_{joint}^{-}) were calculated as the sum of all positive or negative net joint power integrated over time, respectively (Schache et al., 2015). Stance phase joint extensor moment angular impulse (I_{joint}) was calculated as the sum of all stance phase extensor (plantar-flexor for ankle) joint moment integrated with time (DeVita et al., 2000; Winter et al., 1990). Total lower extremity support torque (I_{total}) was calculated as the sum of ankle, knee and hip joint stance phase extensor moment angular impulse (DeVita et al., 2000; Winter et al., 1990), expressed as: I_{total} = I_{ankle} + I_{knee} + I_{hip}. Joint level kinematic variables were calculated from the output results in Visual 3D. Joint ground contact angle (θ_{joint}^{GCA}) and toe-off angle (θ_{joint}^{TOA}) were chosen from the joint angle values in the first frame and the last frame of stance phase. Joint peak extension (plantar flexion for ankle) angle (θ_{joint}^{PEA}) and joint peak flexion (dorsiflexion for ankle) angle (θ_{joint}^{PFA}) were chosen from the maximum extension and flexion joint angle in a whole gait cycle, respectively. Joint angle range of motion (θ_{joint}^{ROM}) was calculated as the difference between peak flexion angle and peak extension angle within the gait cycle. In this study, the sagittal plane neutral position of each join was defined as the zero-degree angle, joint flexion (dorsiflexion for ankle) as negative and joint extension (plantarflexion for ankle) as positive, compared with joint neutral position. The sagittal plane moment and net joint power calculation for each joint shared the same principle.

General gait variables (step length, step width, stance time, swing time and gait cadence) were also calculated. Step length was calculated as the distance between two consecutive heel strikes, step width was calculated as the distance between the outer most borders of two consecutive steps (Brach, Berlin, VanSwearingen, Newman, & Studenski, 2005). In this study we used the calcaneus marker and lateral malleolus marker positions to estimate step length and step width, respectively. Stance phase time was determined as the time when the foot was in contact with the treadmill and swing phase time was determined as the time when the foot was not in contact with the treadmill, based upon the ground reaction force value (Brach et al., 2005).

Group average net joint power curve was plotted over a full gait cycle. We chose three different representative walking (1.0, 1.4, 1.8 m/s) and running (1.8, 2.6, 3.8 m/s) speeds, to show joint kinetic characteristics during slow, medium and relatively fast locomotion speed conditions. Stance phase sagittal plane ankle joint angle and moment values were resampled and averaged within each group and plotted to examine the ankle joint dynamic loading response and work performed in different walking and running speeds.

All outcome variables were calculated and averaged from both limbs, normalized to body mass (where appropriate) and averaged across three gait cycles.

Statistical Analysis

Joint stiffness (K_{joint}), joint work (W_{joint}), angular impulse (I_{joint}) and all joint kinematic variables were compared in a 2-way mixed effects ANOVAs (group × speed) for each joint, within walking and running conditions, respectively in SPSS (V22.0, IBM, Armonk, NY). The factor of group (young vs. middle age group) was tested for between subject effect and speed was tested for within subject effect in the statistical analysis.

Initial alpha level was set to 0.05. When main effect or interaction effect were detected, Bonferroni adjustments were used for pairwise comparison. Follow up pairwise comparison alpha level was set to 0.05 divided by the number of comparisons. An unpaired sample t-test was conducted to test general gait pattern variables (step length, step width, stance time, swing time and gait cadence) in each speed between young and middle age group.

Results

In both walking and running conditions, there was no significant difference between young and middle age subjects in K_{ankle}, K_{knee} and K_{hip} (Table 4.1).

Table 4.1. Joint stiffness (Nm/kg/deg) between the young age group (n = 10) and the middle age group (n = 10) across walking and running speeds. Sample Mean (SD).

	Ankle		Knee		Hip	
	Young	Middle	Young	Middle	Young	Middle
Walk						
0.8 m/s	0.11 (0.09)	0.11 (0.05)	0.13 (0.05)	0.13 (0.07)	0.06 (0.03)	0.06 (0.07)
1.0 m/s	0.11 (0.07)	0.16 (0.08)	0.13 (0.04)	0.13 (0.05)	0.07 (0.03)	0.07 (0.02)
1.2 m/s	0.08 (0.04)	0.11 (0.06)	0.08 (0.01)	0.09 (0.02)	0.06 (0.01)	0.06 (0.02)
1.4 m/s	0.09 (0.03)	0.10 (0.06)	0.08 (0.02)	0.08 (0.02)	0.06 (0.01)	0.06 (0.02)
1.6 m/s	0.07 (0.02)	0.13 (0.07)	0.08 (0.02)	0.10 (0.04)	0.07 (0.02)	0.07 (0.03)
1.8 m/s	0.09 (0.02)	0.12 (0.05)	0.09 (0.01)	0.10 (0.03)	0.07 (0.01)	0.08 (0.03)
2.0 m/s	0.09 (0.02)	0.13 (0.04)	0.11 (0.02)	0.12 (0.04)	0.09 (0.02)	0.10 (0.05)
Run						
1.8 m/s	0.15 (0.05)	0.22 (0.10)	0.09 (0.02)	0.11 (0.03)	0.25 (0.11)	0.26 (0.17)
2.2 m/s	0.17 (0.05)	0.19 (0.04)	0.10 (0.03)	0.11 (0.02)	0.26 (0.15)	0.19 (0.04)
2.6 m/s	0.18 (0.08)	0.20 (0.04)	0.11 (0.03)	0.13 (0.03)	0.33 (0.14)	0.19 (0.04)
3.0 m/s	0.19 (0.05)	0.20 (0.11)	0.13 (0.03)	0.14 (0.04)	0.28 (0.06)	0.19 (0.05)
3.4 m/s	0.22 (0.09)	0.20 (0.05)	0.16 (0.05)	0.15 (0.06)	0.34 (0.08)	0.19 (0.08)
3.8 m/s	0.25 (0.07)	0.21 (0.10)	0.16 (0.05)	0.20 (0.09)	0.33 (0.10)	0.23 (0.07)

For W_{joint} in walking and running conditions, generally there were not much difference between young and middle age group for each joint. During the stance phase of walking, W_{hip}^{+} for the middle age group was higher than for the young age group (*p* = .029) across all walking speeds (except at 0.8 m/s) (Table 4.2). In the stance phase of walking, W_{hip}^{-} for the middle age group was lower than for the young age group (*p* = .031) across all walking speeds (except at 0.8 m/s) (Table 4.2).

In the stance phase of walking, the middle age group had higher ankle joint plantar flexor moment angular impulse (I_{ankle}) compared with the young age group (*p* = .002) across all walking speeds (Table 4.4). The middle age group also had a higher total support torque (I_{total}) compared with the young age group (*p* = .016) across all walking speeds. However, in the stance phase of running, there were no differences between young and middle age group for I_{joint} and I_{total}.

Table 4.2. Joint work (J/kg) between the young age group (n = 10) and the middle age group (n = 10) in stance and swing phase across walking speeds. Sample Mean (SD).

	Ankle		Knee		Hip*	
	Young	Middle	Young	Middle	Young	Middle
Stance Phase Positive Work						
0.8 m/s	0.13 (0.06)	0.12 (0.06)	0.10 (0.07)	0.05 (0.03)	0.09 (0.05)	0.08 (0.03)
1.0 m/s	0.18 (0.08)	0.18 (0.03)	0.10 (0.07)	0.10 (0.06)	**0.13 (0.05)**	**0.15 (0.04)**
1.2 m/s	0.18 (0.08)	0.28 (0.18)	0.11 (0.04)	0.14 (0.07)	**0.09 (0.04)**	**0.14 (0.07)**
1.4 m/s	0.25 (0.07)	0.23 (0.09)	0.13 (0.03)	0.11 (0.04)	**0.12 (0.03)**	**0.16 (0.06)**
1.6 m/s	0.27 (0.12)	0.38 (0.18)	0.18 (0.07)	0.23 (0.15)	**0.13 (0.05)**	**0.20 (0.08)**
1.8 m/s	0.31 (0.12)	0.41 (0.20)	0.22 (0.10)	0.21 (0.09)	**0.16 (0.07)**	**0.20 (0.06)**
2.0 m/s	0.34 (0.16)	0.45 (0.17)	0.22 (0.14)	0.22 (0.11)	**0.18 (0.08)**	**0.27 (0.08)**
Stance Phase Negative Work						
0.8 m/s	0.24 (0.07)	0.16 (0.05)	0.07 (0.06)	0.13 (0.08)	0.07 (0.06)	0.09 (0.07)
1.0 m/s	0.22 (0.05)	0.21 (0.05)	0.14 (0.14)	0.19 (0.12)	**0.11 (0.13)**	**0.10 (0.06)**
1.2 m/s	0.16 (0.09)	0.21 (0.11)	0.21 (0.18)	0.15 (0.03)	**0.23 (0.20)**	**0.10 (0.05)**
1.4 m/s	0.18 (0.05)	0.18 (0.04)	0.15 (0.03)	0.18 (0.10)	**0.13 (0.07)**	**0.12 (0.14)**
1.6 m/s	0.13 (0.05)	0.24 (0.14)	0.27 (0.24)	0.30 (0.24)	**0.32 (0.35)**	**0.16 (0.06)**
1.8 m/s	0.13 (0.05)	0.16 (0.10)	0.33 (0.26)	0.24 (0.11)	**0.36 (0.41)**	**0.17 (0.09)**
2.0 m/s	0.11 (0.03)	0.14 (0.16)	0.24 (0.08)	0.29 (0.17)	**0.22 (0.06)**	**0.16 (0.06)**
Swing Phase Positive Work						
0.8 m/s	<0.01 (0.00)	<0.01 (0.00)	0.01 (0.01)	<0.01 (0.01)	0.04 (0.02)	0.04 (0.02)
1.0 m/s	<0.01 (0.00)	<0.01 (0.00)	<0.01 (0.00)	0.01 (0.00)	0.06 (0.03)	0.05 (0.01)
1.2 m/s	<0.01 (0.00)	<0.01 (0.00)	<0.01 (0.00)	0.01 (0.02)	0.08 (0.03)	0.07 (0.02)
1.4 m/s	0.01 (0.00)	0.01 (0.00)	<0.01 (0.00)	0.01 (0.00)	0.08 (0.01)	0.09 (0.02)
1.6 m/s	0.01 (0.00)	0.01 (0.00)	0.01 (0.01)	0.01 (0.01)	0.10 (0.03)	0.10 (0.03)
1.8 m/s	0.01 (0.00)	0.01 (0.00)	0.01 (0.01)	0.01 (0.01)	0.13 (0.03)	0.12 (0.03)
2.0 m/s	0.01 (0.00)	0.01 (0.00)	0.01 (0.01)	0.01 (0.01)	0.16 (0.04)	0.16 (0.04)

Swing Phase Negative Work						
0.8 m/s	<0.01 (0.00)	<0.01 (0.00)	0.10 (0.02)	0.10 (0.02)	<0.01 (0.01)	<0.01 (0.00)
1.0 m/s	<0.01 (0.00)	<0.01 (0.00)	0.13 (0.02)	0.13 (0.02)	<0.01 (0.00)	<0.01 (0.00)
1.2 m/s	<0.01 (0.00)	<0.01 (0.00)	0.16 (0.02)	0.15 (0.02)	<0.01 (0.00)	<0.01 (0.00)
1.4 m/s	<0.01 (0.00)	<0.01 (0.00)	0.17 (0.01)	0.16 (0.03)	<0.01 (0.00)	0.01 (0.00)
1.6 m/s	<0.01 (0.00)	<0.01 (0.00)	0.18 (0.04)	0.19 (0.03)	0.01 (0.00)	0.01 (0.01)
1.8 m/s	<0.01 (0.00)	<0.01 (0.00)	0.22 (0.03)	0.22 (0.02)	0.01 (0.01)	0.02 (0.01)
2.0 m/s	<0.01 (0.00)	<0.01 (0.00)	0.26 (0.03)	0.27 (0.03)	0.02 (0.02)	0.03 (0.01)

Note: <0.01 indicates a negligible value; Joint negative work data were presented in absolute values.
*Statistically significant difference between young and middle age groups across speeds, is indicated in bold.

Table 4.3. Joint work (J/kg) between the young age group (n = 10) and the middle age group (n = 10) in stance and swing phase across running speeds. Sample Mean (SD).

	Ankle		Knee		Hip	
	Young	Middle	Young	Middle	Young	Middle
Stance Phase Positive Work						
1.8 m/s	0.46 (0.18)	0.49 (0.12)	0.20 (0.06)	0.23 (0.05)	0.04 (0.02)	0.05 (0.05)
2.2 m/s	0.51 (0.21)	0.49 (0.15)	0.23 (0.05)	0.25 (0.06)	0.05 (0.03)	0.09 (0.06)
2.6 m/s	0.46 (0.26)	0.56 (0.11)	0.24 (0.06)	0.25 (0.07)	0.13 (0.16)	0.10 (0.07)
3.0 m/s	0.50 (0.24)	0.52 (0.11)	0.30 (0.11)	0.25 (0.06)	0.14 (0.08)	0.16 (0.10)
3.4 m/s	0.52 (0.28)	0.55 (0.20)	0.29 (0.13)	0.29 (0.13)	0.21 (0.13)	0.19 (0.11)
3.8 m/s	0.70 (0.17)	0.62 (0.19)	0.30 (0.10)	0.23 (0.09)	0.21 (0.05)	0.24 (0.16)
Stance Phase Negative Work						
1.8 m/s	0.30 (0.09)	0.32 (0.06)	0.37 (0.17)	0.29 (0.09)	0.11 (0.07)	0.17 (0.06)
2.2 m/s	0.31 (0.10)	0.34 (0.08)	0.38 (0.13)	0.29 (0.13)	0.12 (0.07)	0.19 (0.08)
2.6 m/s	0.31 (0.09)	0.37 (0.09)	0.42 (0.22)	0.33 (0.12)	0.13 (0.11)	0.22 (0.11)
3.0 m/s	0.35 (0.14)	0.39 (0.12)	0.46 (0.19)	0.25 (0.13)	0.18 (0.11)	0.26 (0.13)
3.4 m/s	0.33 (0.14)	0.41 (0.19)	0.45 (0.14)	0.30 (0.14)	0.20 (0.16)	0.27 (0.15)
3.8 m/s	0.44 (0.07)	0.46 (0.18)	0.36 (0.05)	0.26 (0.10)	0.22 (0.17)	0.31 (0.14)
Swing Phase Positive Work						
1.8 m/s	0.01 (0.00)	0.01 (0.00)	0.01 (0.01)	0.01 (0.01)	0.15 (0.02)	0.18 (0.04)
2.2 m/s	0.01 (0.00)	0.01 (0.00)	0.01 (0.01)	0.01 (0.01)	0.22 (0.05)	0.25 (0.05)
2.6 m/s	0.01 (0.00)	0.01 (0.00)	0.02 (0.01)	0.01 (0.01)	0.32 (0.06)	0.33 (0.08)
3.0 m/s	0.01 (0.00)	0.01 (0.00)	0.02 (0.01)	0.01 (0.01)	0.44 (0.11)	0.45 (0.07)
3.4 m/s	0.01 (0.00)	0.02 (0.01)	0.02 (0.02)	0.01 (0.01)	0.56 (0.12)	0.53 (0.14)
3.8 m/s	0.02 (0.00)	0.02 (0.00)	0.03 (0.03)	0.01 (0.01)	0.67 (0.14)	0.72 (0.16)
Swing Phase Negative Work						
1.8 m/s	<0.01 (0.00)	<0.01 (0.00)	0.26 (0.03)	0.27 (0.03)	0.01 (0.01)	0.01 (0.01)

2.2 m/s	<0.01 (0.00)	<0.01 (0.00)	0.35 (0.03)	0.38 (0.05)	0.02 (0.02)	0.02 (0.01)
2.6 m/s	<0.01 (0.00)	<0.01 (0.00)	0.47 (0.06)	0.48 (0.09)	0.04 (0.02)	0.03 (0.02)
3.0 m/s	<0.01 (0.00)	0.01 (0.00)	0.58 (0.07)	0.62 (0.10)	0.05 (0.02)	0.06 (0.04)
3.4 m/s	0.01 (0.00)	0.01 (0.00)	0.73 (0.08)	0.73 (0.20)	0.08 (0.03)	0.09 (0.04)
3.8 m/s	0.01 (0.00)	0.01 (0.00)	0.88 (0.14)	0.88 (0.18)	0.08 (0.04)	0.11 (0.05)

Note: <0.01 indicates a negligible value; Joint negative work data were presented in absolute values.

Table 4.4. Stance phase joint extensor moment angular impulse (Nm·s/kg) and total support torque (Nm·s/kg) between young age group (n = 10) and middle age group (n = 10) across walking and running speeds. Sample Mean (SD).

	Ankle*		Knee		Hip		Total*	
	Young	Middle	Young	Middle	Young	Middle	Young	Middle
Walk								
0.8 m/s	**0.53 (0.15)**	**0.53 (0.18)**	0.04 (0.05)	0.12 (0.08)	0.07 (0.04)	0.04 (0.03)	**0.65 (0.20)**	**0.67 (0.15)**
1.0 m/s	**0.43 (0.10)**	**0.52 (0.09)**	0.09 (0.12)	0.14 (0.11)	0.09 (0.05)	0.09 (0.02)	**0.61 (0.11)**	**0.75 (0.12)**
1.2 m/s	**0.36 (0.17)**	**0.54 (0.16)**	0.14 (0.14)	0.11 (0.08)	0.06 (0.04)	0.09 (0.04)	**0.56 (0.10)**	**0.74 (0.14)**
1.4 m/s	**0.39 (0.07)**	**0.43 (0.11)**	0.10 (0.05)	0.09 (0.08)	0.09 (0.02)	0.11 (0.06)	**0.59 (0.06)**	**0.63 (0.08)**
1.6 m/s	**0.30 (0.12)**	**0.54 (0.24)**	0.17 (0.16)	0.08 (0.03)	0.10 (0.05)	0.12 (0.05)	**0.56 (0.09)**	**0.69 (0.22)**
1.8 m/s	**0.30 (0.10)**	**0.46 (0.20)**	0.21 (0.22)	0.13 (0.08)	0.11 (0.03)	0.11 (0.03)	**0.62 (0.23)**	**0.70 (0.25)**
2.0 m/s	**0.29 (0.06)**	**0.38 (0.12)**	0.13 (0.06)	0.18 (0.18)	0.12 (0.04)	0.14 (0.02)	**0.53 (0.09)**	**0.69 (0.30)**
Run								
1.8 m/s	0.34 (0.08)	0.38 (0.06)	0.21 (0.06)	0.23 (0.06)	0.05 (0.01)	0.05 (0.04)	0.60 (0.06)	0.65 (0.09)
2.2 m/s	0.30 (0.12)	0.35 (0.05)	0.21 (0.04)	0.21 (0.06)	0.06 (0.02)	0.07 (0.03)	0.58 (0.14)	0.63 (0.08)
2.6 m/s	0.28 (0.12)	0.36 (0.04)	0.20 (0.08)	0.21 (0.08)	0.08 (0.06)	0.07 (0.03)	0.56 (0.11)	0.64 (0.07)
3.0 m/s	0.29 (0.12)	0.34 (0.05)	0.21 (0.05)	0.18 (0.07)	0.09 (0.04)	0.08 (0.03)	0.60 (0.11)	0.60 (0.09)
3.4 m/s	0.26 (0.13)	0.33 (0.06)	0.21 (0.05)	0.20 (0.13)	0.10 (0.06)	0.08 (0.04)	0.57 (0.11)	0.60 (0.10)

3.8 m/s	0.35 (0.06)	0.34 (0.05)	0.17 (0.04)	0.17 (0.05)	0.10 (0.03)	0.08 (0.04)	0.61 (0.06)	0.59 (0.06)

*Statistically significant difference between young and middle age groups across speeds are indicated in bold.

For joint angle comparison between two groups, there were no differences between young and middle age group at either the ankle or hip joint, however there were some differences at the knee joint. In the walking condition, middle age group had a higher knee flexion angle at ground contact (θ_{knee}^{GCA}) (p = .005) and toe off (θ_{knee}^{TOA}) (p < .001) across all walking speeds (Table 4.5). For θ_{knee}^{PEA} in walking, the young age group had a higher knee extension angle over the whole gait cycle across all walking speeds (p = .003) (Table 4.5). In running condition, the middle age group had a higher knee flexion angle at ground contact (θ_{knee}^{GCA}) compared with the young age group (p = .037) (Table 4.5). Similar to the walking condition the young age group had a higher θ_{knee}^{PEA} compared with the middle age group (p = .039) across all running speeds (Table 4.5).

For general comparison of gait variables between the two groups, generally there was no difference in step length, step width, stance time, swing time and gait cadence in all walking speeds. The only difference was that swing phase time in the young age group was longer than in the middle age group (p = .03) at speed 2.0 m/s (Table 4.6). In running condition, the young age group had longer swing phase time in all speeds between 2.6 – 3.8 m/s (p = .01, p = .03, p = .004, p = .04 respectively) (Table 4.6). The middle age group had higher gait cadence at speed 3.0 m/s (p = .009) and 3.4 m/s (p = .03) (Table 4.6).

Among three representative walking speeds (1.0, 1.4, 1.8 m/s) and running speeds (1.8, 2.6, 3.8 m/s), net joint power at all three joints was very similar between the young and middle age groups (Fig. 4.1, Fig. 4.2). Lastly, the magnitude of the flexion (dorsiflexion for ankle) and extension (plantar-flexion for ankle) power values tended to increase when locomotion speeds increased (Fig. 4.1, Fig. 4.2).

Table 4.5. Joint angle (degree) between the young age group (n = 10) and the middle age group (n = 10) across walking and running speeds. Sample Mean (SD).

	GCA*		TOA*		PFA		PEA*		ROM	
	Young	Middle	Young	Middle	Young	Middle	Young	Middle	Young	Middle
Ankle										
Walk										
0.8 m/s	11.19 (2.73)	8.12 (4.37)	11.59 (5.46)	12.44 (4.83)	-1.60 (2.04)	-2.77 (3.50)	20.85 (5.27)	17.34 (4.59)	22.45 (5.34)	20.11 (3.77)
1.0 m/s	9.09 (2.50)	6.93 (3.44)	15.27 (4.45)	15.21 (5.20)	-2.03 (3.50)	-3.82 (5.11)	24.26 (5.81)	20.40 (5.89)	26.29 (5.24)	24.21 (3.49)
1.2 m/s	8.56 (2.70)	6.89 (2.47)	14.13 (7.72)	13.73 (6.72)	1.62 (3.57)	3.44 (2.33)	24.22 (4.67)	20.72 (4.88)	25.84 (4.39)	24.16 (4.21)
1.4 m/s	6.75 (2.78)	6.29 (3.18)	19.60 (4.96)	15.91 (5.35)	-1.73 (4.23)	-2.37 (3.65)	26.39 (5.10)	22.37 (5.04)	28.12 (4.52)	24.75 (4.27)
1.6 m/s	6.31 (2.95)	4.62 (2.92)	19.69 (4.76)	15.24 (4.46)	-0.69 (3.10)	-2.45 (3.88)	26.90 (4.26)	22.37 (5.40)	27.60 (4.65)	24.83 (4.17)
1.8 m/s	5.30 (2.73)	3.89 (2.62)	21.18 (5.96)	18.37 (5.16)	0.52 (2.22)	-1.93 (3.41)	28.01 (5.27)	23.34 (5.06)	27.49 (5.12)	25.27 (3.87)
2.0 m/s	2.98 (3.09)	3.09 (3.02)	21.42 (6.18)	20.62 (4.26)	-0.43 (2.77)	-2.11 (7.10)	27.09 (3.81)	23.80 (5.33)	27.52 (4.91)	25.91 (3.47)
Run										
1.8 m/s	5.22 (5.85)	0.69 (3.73)	21.40 (7.01)	16.82 (4.95)	-9.42 (2.78)	-12.89 (3.51)	26.63 (5.46)	21.02 (5.26)	36.05 (3.49)	33.91 (4.35)
2.2 m/s	3.42 (5.22)	1.51 (2.99)	18.98 (10.84)	18.23 (5.79)	-11.21 (3.65)	-12.49 (3.36)	28.49 (6.24)	24.95 (6.88)	39.70 (3.60)	37.44 (6.69)
2.6 m/s	3.45 (4.69)	1.42 (3.39)	21.30 (6.72)	19.49 (6.14)	-10.28 (3.16)	-12.22 (3.58)	30.30 (5.02)	27.42 (7.64)	40.58 (2.94)	39.64 (7.25)
3.0 m/s	3.09 (5.03)	2.81 (4.29)	18.22 (6.47)	16.86 (6.91)	-11.17 (2.57)	-11.94 (3.47)	31.97 (4.66)	28.01 (5.66)	43.14 (4.76)	39.95 (5.53)
3.4 m/s	2.33 (5.91)	2.66 (5.69)	16.00 (11.06)	16.08 (7.34)	-11.60 (4.11)	-11.96 (4.54)	31.60 (4.67)	29.42 (5.51)	43.20 (4.88)	41.37 (5.40)

3.8 m/s	2.35 (4.32)	4.98 (5.58)	20.72 (5.06)	18.10 (5.09)	-10.34 (3.84)	-11.26 (4.44)	32.35 (4.20)	31.39 (7.01)	42.69 (4.80)	42.65 (7.63)
Knee										
Walk										
0.8 m/s	**-0.88** **(4.40)**	**-7.95** **(4.56)**	**-38.92** **(6.86)**	**-44.07** **(7.53)**	-65.04 (4.18)	-65.37 (3.87)	**1.22** **(4.06)**	**-3.91** **(3.26)**	66.26 (4.19)	61.46 (3.78)
1.0 m/s	**0.85** **(4.66)**	**-7.07** **(6.23)**	**-41.35** **(8.42)**	**-48.09** **(6.59)**	-67.74 (3.65)	-70.26 (6.95)	**2.82** **(4.98)**	**-4.30** **(5.74)**	70.55 (3.74)	65.96 (2.53)
1.2 m/s	**-0.18** **(4.12)**	**-6.41** **(5.14)**	**-36.21** **(9.47)**	**-43.98** **(11.51)**	-67.17 (3.91)	-70.04 (3.30)	**1.70** **(3.72)**	**-2.82** **(2.48)**	68.87 (3.14)	67.22 (2.91)
1.4 m/s	**-0.53** **(4.60)**	**-7.16** **(4.62)**	**-40.37** **(3.95)**	**-40.95** **(6.22)**	-67.32 (5.61)	-68.81 (4.29)	**1.43** **(3.90)**	**-2.87** **(2.38)**	68.75 (3.24)	65.94 (4.02)
1.6 m/s	**-3.75** **(4.28)**	**-8.11** **(4.47)**	**-35.02** **(6.46)**	**-43.19** **(11.32)**	-65.37 (5.59)	-70.65 (7.38)	**2.25** **(4.28)**	**-3.41** **(4.99)**	67.62 (3.26)	67.24 (4.43)
1.8 m/s	**-3.83** **(4.23)**	**-6.92** **(3.65)**	**-33.71** **(7.46)**	**-41.62** **(7.25)**	-64.83 (5.62)	-68.88 (5.46)	**1.75** **(4.07)**	**-2.15** **(2.78)**	66.58 (4.49)	66.74 (5.09)
2.0 m/s	**-6.07** **(4.96)**	**-9.36** **(6.55)**	**-33.99** **(9.84)**	**-46.41** **(5.39)**	-64.97 (5.43)	-71.31 (9.01)	**1.76** **(4.19)**	**-3.90** **(6.66)**	66.73 (4.12)	67.42 (5.21)
Run										
1.8 m/s	**-9.94** **(5.05)**	**-18.03** **(5.19)**	-18.35 (7.38)	-25.46 (8.80)	-77.44 (7.85)	-85.07 (11.81)	**-7.24** **(5.56)**	**-14.65** **(5.52)**	70.20 (8.12)	70.42 (11.93)
2.2 m/s	**-12.69** **(4.09)**	**-17.23** **(5.79)**	-17.58 (5.69)	-20.52 (6.58)	-87.58 (9.94)	-87.47 (12.20)	**-8.66** **(3.63)**	**-13.97** **(5.87)**	78.92 (10.74)	73.50 (12.27)
2.6 m/s	**-12.86** **(4.61)**	**-17.77** **(5.14)**	-16.96 (7.12)	-18.82 (6.48)	-95.67 (9.72)	-90.92 (11.99)	**-8.62** **(4.61)**	**-12.82** **(5.97)**	87.06 (11.86)	78.10 (11.97)
3.0 m/s	**-13.59** **(6.23)**	**-19.07** **(6.41)**	-15.29 (6.43)	-17.83 (5.93)	-104.06 (11.75)	-99.89 (11.79)	**-8.94** **(4.99)**	**-14.09** **(6.08)**	95.12 (12.11)	85.80 (10.86)
3.4 m/s	**-17.27** **(6.07)**	**-21.88** **(8.12)**	-18.02 (8.25)	-21.25 (9.56)	-111.86 (9.09)	-104.11 (13.13)	**-10.65** **(4.89)**	**-15.12** **(4.30)**	101.20 (8.91)	88.99 (12.17)
3.8 m/s	**-18.67** **(6.52)**	**-21.91** **(5.15)**	-15.82 (3.48)	-20.01 (4.98)	-120.17 (8.45)	-108.65 (9.20)	**-11.70** **(4.27)**	**-17.07** **(5.26)**	108.47 (7.09)	91.58 (9.96)

Hip										
Walk										
0.8 m/s	0.78 (7.18)	0.90 (9.95)	21.47 (7.83)	20.19 (10.30)	-5.91 (6.61)	-3.19 (10.43)	29.70 (6.96)	30.42 (10.39)	35.61 (3.12)	33.61 (3.95)
1.0 m/s	-1.18 (7.50)	-1.56 (9.01)	22.98 (7.35)	23.32 (9.06)	-7.50 (5.57)	-6.14 (10.02)	30.82 (5.50)	32.99 (10.05)	38.31 (2.71)	39.13 (3.63)
1.2 m/s	-3.29 (7.39)	-2.89 (9.20)	27.92 (7.55)	23.34 (13.27)	-8.06 (6.31)	-7.23 (10.42)	33.86 (5.77)	34.54 (10.26)	41.93 (4.39)	41.77 (3.38)
1.4 m/s	-6.00 (7.41)	-5.20 (9.15)	27.98 (5.70)	28.16 (10.73)	-9.36 (6.29)	-8.05 (9.94)	35.44 (4.94)	35.47 (10.85)	44.80 (3.89)	43.52 (4.49)
1.6 m/s	-8.67 (7.16)	-7.67 (8.66)	31.44 (5.84)	27.75 (15.82)	-10.50 (6.55)	-10.17 (9.67)	37.43 (5.72)	37.98 (10.20)	47.93 (4.24)	48.15 (5.44)
1.8 m/s	-10.83 (7.17)	-8.53 (9.29)	31.71 (7.48)	27.59 (12.62)	-12.94 (6.45)	-10.94 (9.53)	38.00 (4.95)	37.51 (10.98)	50.93 (4.51)	48.45 (4.60)
2.0 m/s	-13.94 (8.50)	-11.00 (9.95)	31.19 (8.04)	26.63 (12.10)	-15.49 (8.27)	-13.80 (10.79)	38.10 (6.60)	37.94 (9.73)	53.59 (4.65)	51.74 (6.27)
Run										
1.8 m/s	-1.01 (7.68)	-1.14 (10.17)	25.45 (5.98)	25.81 (10.28)	-8.10 (8.05)	-7.93 (11.18)	26.22 (6.43)	27.05 (10.53)	34.32 (4.76)	34.98 (4.81)
2.2 m/s	-4.68 (6.94)	-3.94 (10.58)	26.06 (7.01)	29.19 (10.60)	-12.84 (8.10)	-11.07 (12.20)	28.25 (6.44)	29.85 (10.31)	41.09 (6.15)	40.92 (5.33)
2.6 m/s	-5.77 (6.88)	-5.52 (10.72)	27.00 (5.43)	31.47 (12.16)	-16.80 (7.59)	-13.64 (11.31)	29.88 (5.76)	32.30 (11.74)	46.68 (6.06)	45.93 (5.56)
3.0 m/s	-7.69 (7.30)	-7.36 (11.59)	30.43 (5.51)	33.11 (12.21)	-22.10 (8.26)	-18.53 (14.70)	33.47 (5.44)	34.65 (11.56)	55.57 (6.83)	53.18 (6.10)
3.4 m/s	-10.17 (7.91)	-8.15 (12.92)	29.63 (7.18)	33.00 (17.03)	-26.67 (9.70)	-20.17 (15.87)	34.22 (6.03)	37.03 (13.09)	60.88 (8.28)	57.20 (8.68)
3.8 m/s	-11.58 (6.68)	-12.00 (11.63)	31.99 (7.77)	33.74 (11.60)	-31.18 (8.72)	-26.21 (10.86)	34.87 (6.56)	35.55 (11.25)	66.05 (6.66)	61.75 (6.26)

*Statistically significant difference between young and middle age groups across speeds, are indicated in bold.
GCA: joint angle at ground contact; **TOA:** joint angle at toe off; **PFA:** joint peak flexion angle in whole gait cycle;

PEA: joint peak extension angle in whole gait cycle; **ROM:** joint angle range of motion in whole gait cycle.

Table 4.6. Step length (m), step width (m), stance phase time (s), swing phase time (s) and gait cadence (steps/min) between the young age group (n = 10) and the middle age group (n = 10) across walking and running speeds. Sample Mean (SD).

	SL		SW		STT		SWT*		GC*	
	Young	Middle	Young	Middle	Young	Middle	Young	Middle	Young	Middle
Walk										
0.8 m/s	0.56 (0.02)	0.54 (0.02)	0.26 (0.04)	0.26 (0.04)	0.78 (0.05)	0.78 (0.08)	0.45 (0.03)	0.43 (0.06)	97.78 (4.01)	99.65 (4.11)
1.0 m/s	0.62 (0.02)	0.61 (0.03)	0.26 (0.04)	0.25 (0.03)	0.71 (0.03)	0.71 (0.03)	0.41 (0.03)	0.41 (0.03)	107.57 (3.33)	107.80 (4.38)
1.2 m/s	0.67 (0.05)	0.66 (0.04)	0.26 (0.04)	0.26 (0.03)	0.64 (0.04)	0.65 (0.08)	0.40 (0.03)	0.39 (0.07)	114.76 (3.18)	115.83 (6.96)
1.4 m/s	0.74 (0.02)	0.71 (0.04)	0.25 (0.03)	0.25 (0.04)	0.62 (0.03)	0.59 (0.04)	0.38 (0.02)	0.39 (0.05)	120.43 (2.94)	123.00 (7.16)
1.6 m/s	0.78 (0.03)	0.76 (0.05)	0.25 (0.03)	0.24 (0.03)	0.57 (0.04)	0.57 (0.06)	0.36 (0.05)	0.36 (0.06)	128.13 (4.00)	129.51 (7.50)
1.8 m/s	0.82 (0.04)	0.78 (0.05)	0.25 (0.04)	0.25 (0.03)	0.53 (0.04)	0.53 (0.06)	0.36 (0.04)	0.34 (0.04)	135.36 (5.20)	138.29 (8.24)
2.0 m/s	0.83 (0.05)	0.74 (0.14)	0.25 (0.03)	0.24 (0.02)	0.50 (0.03)	0.49 (0.05)	**0.33 (0.03)**	**0.31 (0.03)**	144.64 (6.83)	151.11 (9.77)
Run										
1.8 m/s	0.74 (0.07)	0.72 (0.08)	0.25 (0.04)	0.23 (0.02)	0.31 (0.03)	0.30 (0.05)	0.43 (0.04)	0.43 (0.04)	162.71 (5.70)	164.99 (6.17)
2.2 m/s	0.88 (0.09)	0.86 (0.07)	0.24 (0.04)	0.23 (0.02)	0.27 (0.04)	0.28 (0.03)	0.45 (0.06)	0.42 (0.03)	166.63 (8.74)	171.02 (5.92)
2.6 m/s	1.00 (0.11)	0.97 (0.10)	0.23 (0.03)	0.23 (0.01)	0.23 (0.03)	0.26 (0.03)	**0.47 (0.04)**	**0.42 (0.03)**	170.32 (9.25)	175.80 (4.52)
3.0 m/s	1.13 (0.11)	1.06 (0.08)	0.22 (0.03)	0.22 (0.02)	0.23 (0.02)	0.23 (0.02)	**0.47 (0.05)**	**0.43 (0.03)**	**172.34 (9.01)**	**182.57 (6.37)**
3.4 m/s	1.22 (0.12)	1.14 (0.14)	0.22 (0.03)	0.22 (0.02)	0.21 (0.02)	0.22 (0.03)	**0.47 (0.03)**	**0.42 (0.03)**	**178.67 (8.53)**	**187.15 (7.60)**

3.8 m/s	1.29 (0.13)	1.23 (0.14)	0.22 (0.03)	0.21 (0.01)	0.20 (0.01)	0.20 (0.03)	**0.45 (0.03)**	**0.42 (0.03)**	185.06 (8.74)	194.83 (10.15)

*Statistically significant difference between young and middle age groups at each speed are indicated in bold.

SL: step length; **SW:** step width; **STT:** stance phase time; **SWT:** swing phase time; **GC:** gait cadence.

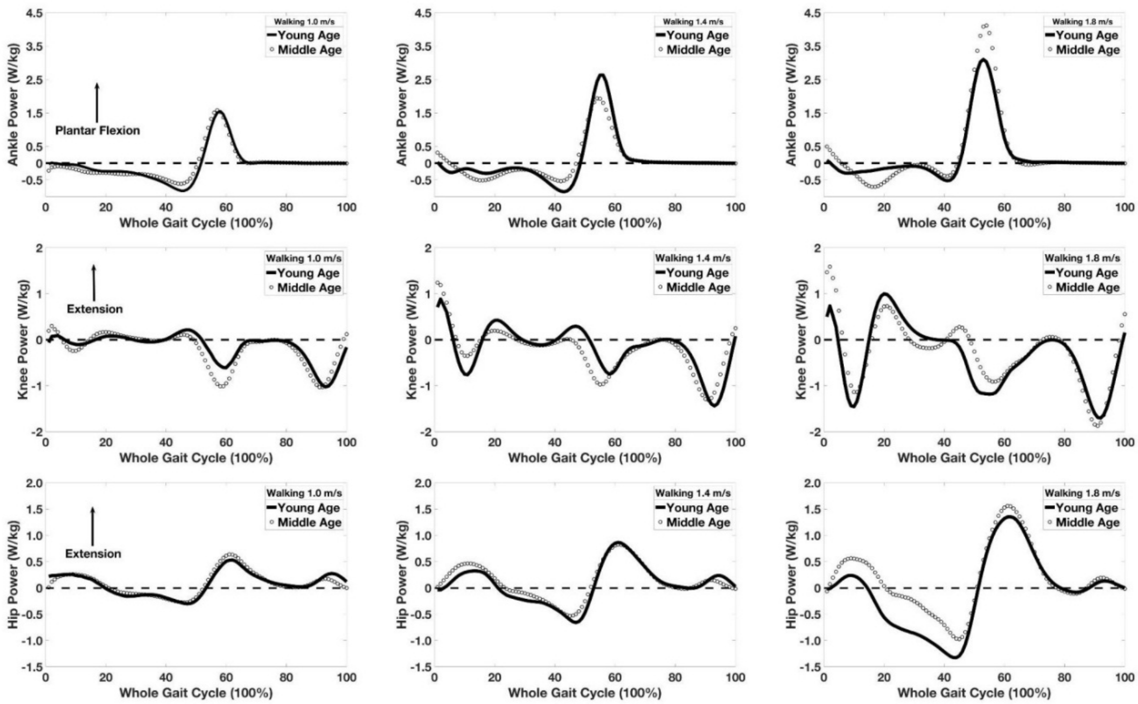

Fig 4.1. Group average ankle, knee and hip joint power curves between the young age group (n = 10) and the middle age group (n = 10) in three representative walking speeds.

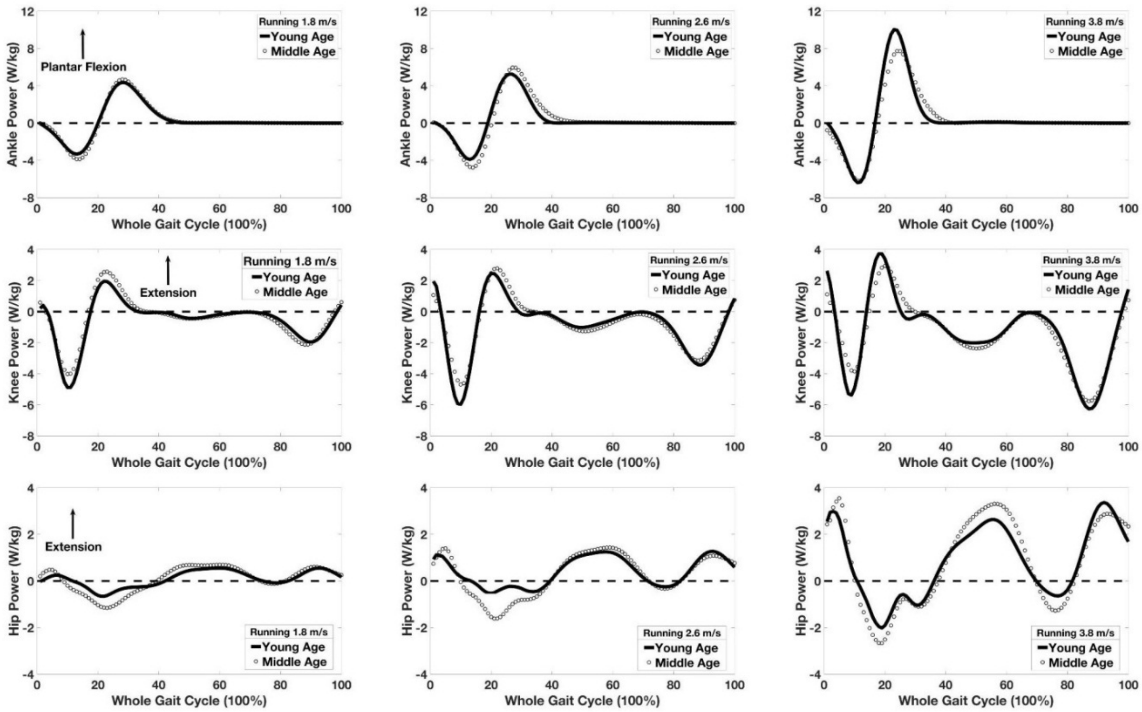

Fig 4.2. Group average ankle, knee and hip joint power curves between the young age group (n = 10) and the middle age group (n = 10) in three representative running speeds.

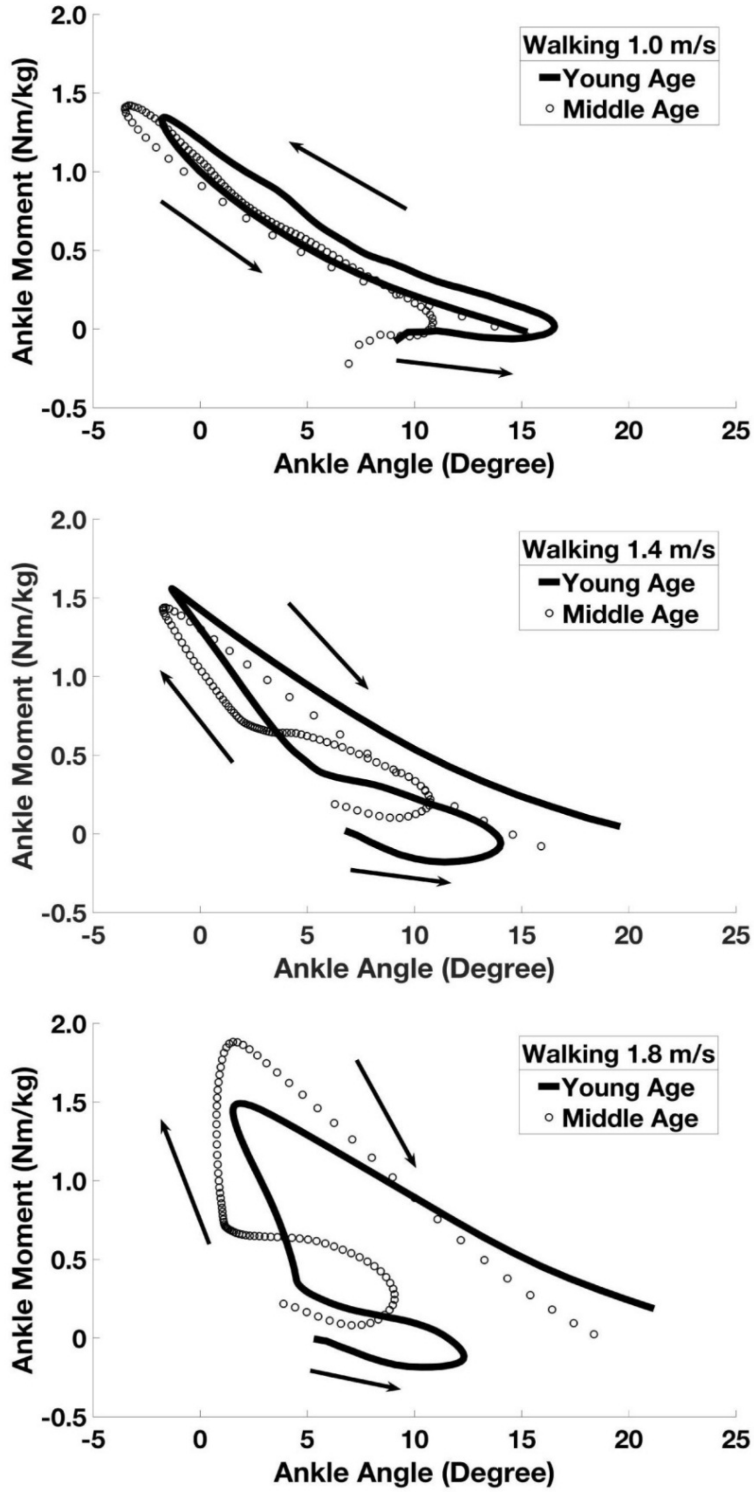

Fig 4.3. Group average ankle joint stance phase angle-moment curves between the young age group (n = 10) and the middle age group (n = 10) in three representative walking speeds.

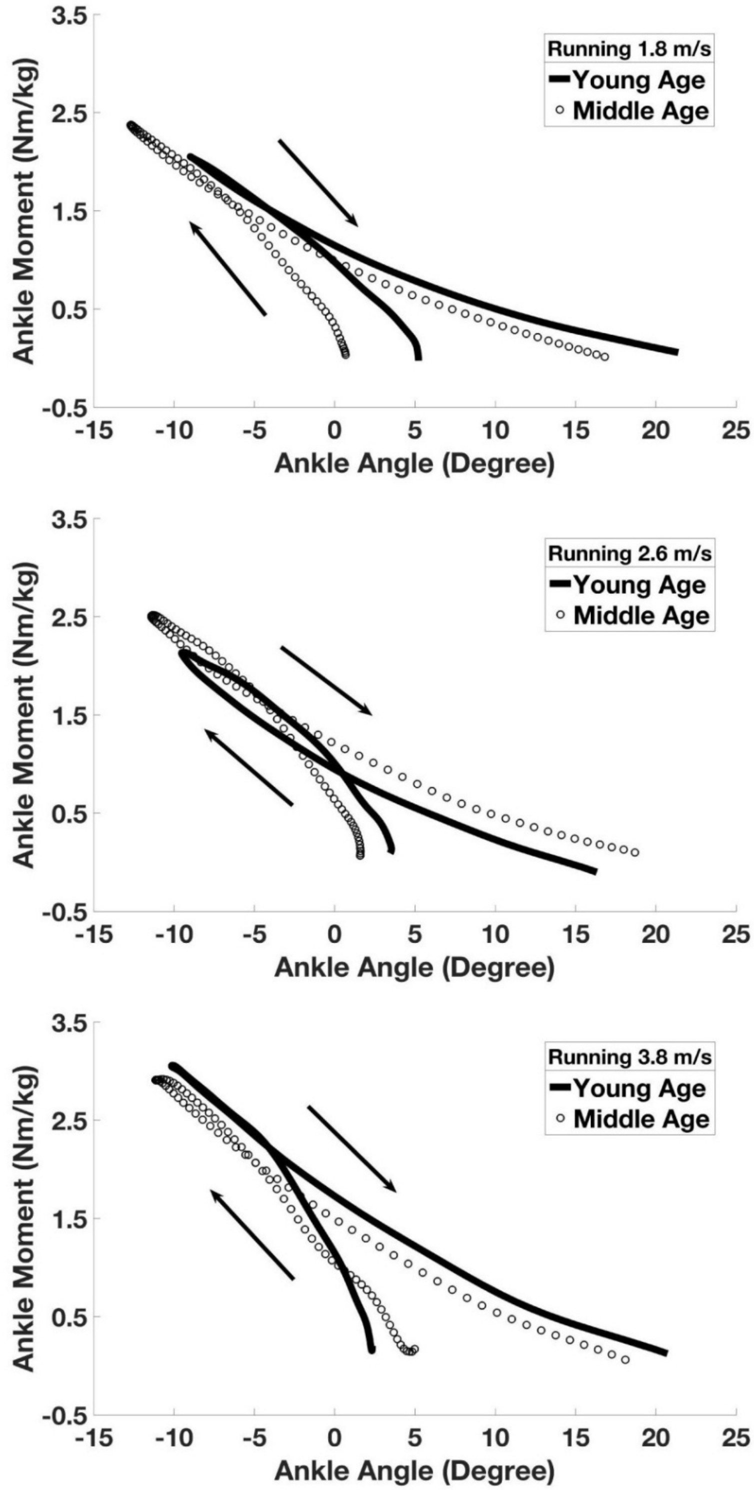

Fig 4.4. Group average ankle joint stance phase angle-moment curves between the young age group (n = 10) and the middle age group (n = 10) in three representative running speeds.

Discussion

The purpose of this research was to investigate lower extremity joint level kinematic and kinetic characteristics, and general gait patterns between young and middle age healthy subjects while walking and running over a range of speeds. The initial hypothesis that the middle age group would have higher joint stiffness was not supported. The hypothesis that the middle age group tend to generate more positive work at the proximal end of lower extremity in walking was supported. Finally, the hypothesis that the middle age group would have higher cadence was partially supported.

For K_{ankle} in walking condition, there was no significant difference between the two groups. At speed 1.0 – 1.2 m/s, and 1.6 – 2.0 m/s, the middle age group was 37%, 32%, 60%, 29% and 37% higher than the young age group, respectively (Table 4.1). In running condition, at speed 1.8 m/s, the middle age group was 38% higher than the young age group. While at speed 3.8 m/s, the middle age group was 17% lower than the young age group (Table 4.1). Similar trends were found for W^{+}_{ankle} in walking stance phase. At speeds 1.2 m/s, and 1.6 – 2.0 m/s, W^{+}_{ankle} for the middle age group was 44%, 34%, 28% and 28% higher compared with the young age group, respectively (Table 4.2). This may indicate that when walking from medium to fast speeds, the middle age group tended to have a higher K_{ankle} value and this would likely be associated with the generation of a higher percentage of stance phase W^{+}_{ankle} as well. Moreover, significantly higher I_{ankle} in the middle age group across all walking speeds would contribute to the higher percentage stance phase W^{+}_{ankle} for the middle age group as well.

In this study, we also investigated ankle joint stance phase angle-moment relationship in both walking and running. We observed a clockwise hysteresis loop in both walking and running. In walking condition, there were 3 apparent phases: initial contact, ascending phase and descending phase (Fig. 4.3) (Crenna & Frigo, 2011). In running condition there were only two phases: an ascending and a descending phase (Fig. 4.4). Subjects in both groups had similar ankle joint stance phase angle-moment relationship, and the patterns agreed with previous studies in both walking and running conditions (Fig. 4.3, Fig. 4.4) (Crenna & Frigo, 2011; Gabriel et al., 2008; Kuitunen et al., 2002). In walking condition, within the initial contact and ascending phase, the area below the curve can be regarded as the negative mechanical work absorbed by the ankle joint, based on the integral of moment over range of rotation principle in the sagittal plane (Crenna & Frigo, 2011). And for the descending phase, the area below the curve can be regarded as the positive mechanical work generated by the ankle joint (Crenna & Frigo, 2011).

At the 1.0 m/s walking speed condition, both groups had a very narrow loop and the descending phase curve went below the ascending phase curve. We also observed that at 1.0 m/s, both groups had 20% and 15% higher amount of W^{-}_{ankle} than W^{+}_{ankle}, respectively (Table 4.2). This may indicate that at relatively slow walking speed, during stance phase the ankle joint does not need to generate much energy compared with the amount of energy absorbed for either age group. At the more normal walking speed (1.4 m/s) and at relatively fast walking speed (1.8 m/s), both groups had a “yielding ascending curve” and as the speed increased, the “yielding” pattern became more obvious, especially for the middle age group (Fig. 4.3). It seems that both groups kept ankle joint in the plantar flexed position throughout the stance phase while walking at 1.8 m/s (Fig. 4.3). This may infer that in the mid-stance to late-stance period (right before push-off), the ankle joint tended to dorsiflex less compared with earlier initial loading period for both groups. For the middle age group, the ankle joint tended to be “stiffer” in late middle stance period compared with the young age group at 1.8 m/s (Fig. 4.3). One possible reason may be due to a relatively higher ankle joint plantar flexor moment from middle stance to terminal stance period for the middle age group, which can result in a higher dynamic loading response of ankle joint musculotendinous system. Another possible reason may be related to age, as the ankle joint sagittal plane range of motion decreased in response to the loading effect. Future research leading to the development of foot-ankle assistive devices may need to focus on the middle stance phase “yielding” energy absorption pattern observed in this study, especially for middle age individuals at fast walking speeds. In the descending phase of the angle-moment curve during walking, the young age group tended to have a greater slope at 1.4 m/s (Fig. 4.3), while at 1.8 m/s, the middle age group tended to have a greater slope (Fig. 4.3). A greater slope in the descending phase indicates a higher amount of late stance energy generation. The observed energy generation patterns (area below descending phase of angle-moment curve) agreed with the observed positive ankle joint power curve in each speed (Fig. 4.1, Fig. 4.3), since mechanical work can be calculated as the integral of net joint power integrated over time (Farris & Sawicki, 2012; Schache et al., 2015). In the running condition, both groups tended to have a “narrow loop” angle-moment relationship, and generally there were not much differences between the two groups at three selected speeds (Fig. 4.4). When running speeds increased, the slopes of ascending and descending curves tended to increase as well (Fig. 4.4).

In the stance phase of walking, the middle age group had higher W_{hip}^{+} across all speeds (Table 4.2). This finding was similar to previous studies comparing an older age group and a young age group during walking (DeVita et al., 2000; Silder et al., 2008). Previous studies suggested that elderly people would produce more positive work and extensor moment angular impulse at the hip joint to compensate for less ankle joint positive work and plantar flexor moment angular impulse to achieve similar gait performance (Browne & Franz, 2017; DeVita et al., 2000; Silder et al., 2008). In the present study, the middle age group also had higher I_{ankle} and I_{total} in walking, inferring that middle age individuals may not use the extra W_{hip}^{+} as a compensatory mechanism within the lower extremity during walking. One possible reason may be related to the smaller age range in this study compared with previous studies comparing elderly and young groups (DeVita et al., 2000; Silder et al., 2008). Another reason might be that most of the middle age subjects in this study were generally fit (BMI: 22.96 $\pm$ 2.88 kg/m^2).

Comparison of joint kinematic patterns revealed that the middle age group had higher knee joint flexion angle at ground contact and toe off period in walking, as well as less extension angle over the gait cycle in both walking and running compared with the young age group (Table 4.5). This appears to be associated with the observation that the middle age group had a lower knee extensor moment and higher knee flexion velocity at ground contact and toe-off period (Piazza & Delp, 1996), which would contribute to knee flexion angle that was higher in the middle age group in both stance and swing phase of walking and running.

In the walking condition, there were few spatiotemporal differences between the two groups (Table 4.6). However, when the speeds increased in the running condition, similar to previous studies (Boyer et al., 2012; DeVita et al., 2000), the middle age group tended to have less swing phase time and higher cadence as a compensation to achieve similar gait performance as young age group.

Limitations

One limitation of the study is that we assumed gait symmetry. All the subjects in both groups were healthy. The outcomes of this study may be only generalizable to the healthy young and middle age population. Another limitation is that during treadmill walking and running, we controlled the locomotion speeds and thus some individual variations may have been restricted.

Future Work

Future studies should investigate lower extremity gait mechanics patterns among healthy young, middle and old age groups, and include comparisons with patient populations over a wider range of walking and running speeds. Additionally, it would be interesting to investigate whole body center of mass dynamics among these populations during different locomotion tasks across speeds.

Conclusion

Lower extremity joint stiffness was not different between healthy young and middle age groups when walking and running across speeds. The middle age group had higher ankle plantar flexor moment angular impulse, higher total lower extremity support torque and higher hip joint positive work in the stance phase of walking. In normal and fast walking speeds, both young and middle age groups tended to have a "yielding ascending curve" in the ankle angle-moment plot, and as speeds increased, the "yielding" energy absorption pattern became more obvious, especially for the middle age group. The middle age group also exhibited higher knee flexion angle at ground contact in walking and running. During some running speeds the middle age group had shorter swing phase time and higher cadence. From the findings of this study, it seems that moderate aging has effects on ankle and hip joint kinetic patterns across walking speeds, knee joint kinematic patterns in both walking and running, and some spatiotemporal changes in running speeds.

Bridge

Chapter IV investigated progression of age potential effects on lower extremity joint level kinematic and kinetic patterns in both walking and running across speeds. This study revealed that moderate aging had effects on lower extremity joints kinematic and kinetic characteristics change. Next, Chapter V will investigate lower extremity joint level kinetic patterns in response to gait transition between walking and running, to further expand the knowledge framework on change of locomotion tasks and speeds effects on lower extremity system.

CHAPTER V

LOWER EXTREMITY JOINT KINETIC PATTERNS IN WALK-TO-RUN AND RUN-TO-WALK TRANSITIONS

This Chapter is currently unpublished. Li Jin designed this study, collected the data and analyzed it. Michael E. Hahn provided mentorship activities, including assistance with study design, general oversight of the project, and editing and finalizing of the journal manuscript.

Introduction

Walking and running are both common locomotion activities for human beings. However each activity has different gait characteristics (Lipfert, Günther, Renjewski, Grimmer, & Seyfarth, 2011) and whole body center of mass dynamic patterns (Alexander, 2003; G. A. Cavagna, Heglund, & Taylor, 1977; C T Farley & Ferris, 1998; Mochon & McMahon, 1980; Veerle Segers et al., 2007; Srinivasan & Ruina, 2006; Willems, Cavagna, & Heglund, 1995). When walking at a constantly increasing speed or running at a constantly decreasing speed, spontaneous walk-to-run transition (WRT) or run-to-walk transition (RWT) dependably occurs at a preferred transition speed (PTS) (Raynor et al., 2002). The magnitude of the acceleration and deceleration affects the speed at which gait transition occurs (L. Li, 2000). Moreover, both WRT and RWT activities happen intuitively within a short period (Hanna, A., Abernethy, B., Neal, R. J., Burgess-Limerick, 2000; Raynor et al., 2002; Veerle Segers, Aerts, Lenoir, & De Clercq, 2008; Thorstensson & Roberthson, 1987). There may be different factors integrated together to modulate the change from one locomotion state to the other.

Previous studies have investigated different factors to explain the gait transition mechanisms for human beings. Generally there have been four proposed mechanisms which modulate and trigger gait transition: metabolic efficiency, mechanical efficiency, mechanical load, cognitive and perceptual modulation (Kung et al., 2018). In regard to energetic cost, walking and running are reported to have relatively low energy cost in the preferred speed range within their own locomotion states (L. Li, 2000; Margaria, Cerretelli, Aghemo, & Sassi, 1963). However, Farris & Sawicki (2012) reported that metabolic cost of transport was higher in running compared to walking at the same speed at 2.0 m/s, and this speed has been reported as a typical adult preferred WRT speed (Raynor et al., 2002). This implies that other mechanisms play important roles in affecting gait transition. With regard to mechanical efficiency, muscle force-length velocity properties (Farris & Sawicki, 2012; Neptune, Clark, & Kautz, 2009), muscle power generation efficiency (Farris & Sawicki, 2012), muscle fiber work and series-elastic element utilization are related to mechanical energy expenditure (Sasaki & Neptune, 2006). Specifically, when walking at speeds above the PTS or running at speeds below PTS, more muscle fiber work is required (Sasaki & Neptune, 2006). There seems to be a feedback system associated with the musculoskeletal system (Farris & Sawicki, 2012) to help minimize mechanical cost of locomotion (Diedrich & Warren, 1995; C T Farley & Ferris, 1998; Kung et al., 2018; Minetti, Ardigo, & Saibene, 1994), which triggers gait transition. Mechanical load trigger, also proposed as the “effort-load hypothesis” mechanism (Pires et al., 2014), is known as a protective injury prevention and muscle stress reduction mechanism in the musculoskeletal system (C. Farley & Taylor, 1991; A Hreljac, 1993; Kung et al., 2018). Specifically, when walking above PTS or running below PTS, the perceived over exertion of some muscles (Farris & Sawicki, 2012; A. Hreljac, Arata, Ferber, Mercer, & Row, 2001; Pires et al., 2014), or protective minimizing peak loads and stress in these muscles to reduce injury risk (Alan Hreljac, 1995; Alan Hreljac, Imamura, Escamilla, Edwards, & MacLeod, 2008; Pires et al., 2014) would lead to a change of locomotion state (gait transition). Lastly, cognitive response to the locomotion difficulty would be likely to modulate gait pattern changes to mitigate the stress on the body. Some other perceptual determinants based on the cognitive feedback, and interaction with the surrounding environment would also be integrated in modulating gait transition (Kung et al., 2018). From a biomechanical perspective it seems that mechanical efficiency and the "effort-load hypothesis" mechanism may be more important factors contributing to gait transition between walking and running (Kung et al., 2018; Pires et al., 2014).

Lower extremity joint level kinetic patterns are closely related to musculoskeletal system mechanical efficiency and mechanical load mechanisms, which help to modulate gait transition between walking and running. Previous studies have investigated walking and running joint kinetic characteristics around the transition speed (Farris & Sawicki, 2012; Pires et al., 2014), as well as gait kinematic and kinetic patterns during the transition process (Alan Hreljac, 1995; Alan Hreljac et al., 2008; V. Segers et al., 2013; Veerle Segers et al., 2008, 2007). When walking at PTS, swing phase peak ankle dorsiflexion power, stance phase peak hip power and moment were higher compared with running at PTS (Pires et al., 2014). This indicates a switch from walking to running would likely reduce the effort of ankle and hip muscles. Additionally, lower extremity joint power generation tends to shift from proximal to distal, when running at PTS compared with walking, as it is deemed to be beneficial for positive mechanical work generation by switching gait patterns (Farris & Sawicki, 2012). In the WRT process, lower extremity joint moment and power characteristics during the transition step were reported as being very similar to running gait (V. Segers et al., 2013). Ankle plantar flexion work, knee extension work and power were regarded as important joint kinetic factors which drive WRT (V. Segers et al., 2013).

Previous research has been focused on WRT gait kinematic and kinetic characteristics, and investigation of possible mechanisms which modulate and trigger gait transition. However, little is known about lower extremity stance phase dynamic loading and response (joint stiffness, ankle joint angle-moment relationship) characteristics, along with joint extensor moment support torque and other joint kinetic characteristics (joint work and power) in both stance and swing phase throughout the full gait transition process (steps before transition, transition step, steps after transition). These measures would provide a deeper understanding of ankle, knee and hip joint functional roles during the gait transition process. Moreover, an increased understanding of these characteristics would be beneficial for future foot-ankle assistive device development, which might be designed to meet multiple locomotion tasks, and be better suited gait transition when the locomotion task changes. In this study, we aim to investigate lower extremity joint stiffness, stance phase joint extensor moment angular impulse, stance and swing phase joint work and power characteristics in the WRT and RWT process. We hypothesize that: (1) lower extremity joints stiffness would increase during WRT, and decrease in the

RWT process; (2) joint work, peak power and extensor moment angular impulse would increase during the WRT, and decrease in the RWT process.

Methods

Recruitment

Ten middle age healthy subjects (51 $\pm$ 6.0 years, 173 $\pm$ 11.4 cm, 70 $\pm$ 15.0 kg) participated in the study. All subjects signed informed written consent approved by the university's institutional review board before participation. All subjects self-reported to be free of lower extremity musculoskeletal related injuries which would affect walking and running for the past 6 months before the test.

Study Design and Experimental Protocol

Fifty-five retro-reflective markers were placed on the skin surface on the subjects, based on a previously published whole body marker set (Sawers & Hahn, 2012). Subjects were first asked to complete the WRT protocol: walking on a force-instrumented treadmill (Bertec, Inc., Columbus, OH) at 1.8 m/s for 30 seconds, then the treadmill was constantly accelerated at 0.1 m/s^2 up to 2.4 m/s. Subjects were asked to transition to a running gait whenever they felt ready during the acceleration process. After transitioning to a running gait, they ran at 2.4 m/s for another 30 seconds. Next, subjects completed the RWT protocol: running at 2.4 m/s for 30 seconds, then the treadmill was constantly decelerated at -0.1 m/s^2 down to 1.8 m/s. Subjects were asked to transition to a walking gait whenever they felt ready during the deceleration process. Once they transitioned to a walking gait, they walked at 1.8 m/s for another 30 seconds. Treadmill acceleration and deceleration magnitude for the WRT and RWT protocols were chosen based on previous work by Segers et al. (V. Segers et al., 2006).

Data Collection

Segmental kinematic data were collected at 120 Hz using an 8-camera motion capture system (Motion Analysis Corp., Santa Rosa, CA). Ground reaction force data were collected at 1200 Hz using the force-instrumented treadmill. Kinematic and kinetic data were filtered with a low-pass fourth-order Butterworth filter at 6 Hz and 50 Hz, respectively.

Data Analysis

In this study, all outcome variables calculation and analysis were focused on the two steps before gait transition (S-2, S-1), the transition step (S0) and the two steps after transition (S1, S2) for both WRT and RWT. Lower extremity joint angles, moments and net joint powers were calculated using an inverse dynamics model coded in Visual 3D (C-Motion, Inc., Germantown, MD). Joint stiffness (K_{joint}) was calculated as the change in sagittal plane joint moment (ΔM_{joint}) divided by sagittal plane joint angular displacement ($\Delta\theta_{joint}$) in the braking phase of ground contact, based on the anterior-posterior ground reaction force value (Hobara et al., 2013; Kuitunen et al., 2002), expressed as: K_{joint} = $\Delta M_{joint}/\Delta\theta_{joint}$. Stance and swing phase joint positive work (W_{joint}^{+}) and negative work (W_{joint}^{-}) were calculated as the sum of all positive or negative net joint power integrated over time, respectively (Schache et al., 2015). Stance phase joint extensor moment angular impulse (I_{joint}) was calculated as the sum of all stance phase extensor (plantar flexor for ankle) joint moment integrated with time (DeVita et al., 2000; Winter et al., 1990). Total lower extremity support torque (I_{total}) was calculated as the sum of ankle, knee and hip joint stance phase extensor moment angular impulse (DeVita et al., 2000; Winter et al., 1990), expressed as: $I_{total} = I_{ankle} + I_{knee} + I_{hip}$. Joint stance and swing phase peak extension and flexion power were chosen from the maximum and minimum joint power value in each phase, respectively. Group average net joint power curves were plotted in all five steps of WRT and RWT, respectively. Stance phase sagittal plane ankle joint angle-moment curves were also resampled and averaged for further analysis.

Statistical Analysis

Joint stiffness (K_{joint}), joint work (W_{joint}) and angular impulse (I_{joint}) were examined for differences between joints and steps before, during and after the transition using a 2-way ANOVAs (joint × step) for WRT and RWT in SPSS (V22.0, IBM, Armonk, NY), respectively. Total support torque (I_{total}), joint stance and swing phase peak extension and flexion power were examined using a 1-way ANOVA to compare between the five steps tested during WRT and RWT, respectively. For this analysis, peak joint extension and flexion power analysis was conducted within ankle, knee and hip, separately. Initial alpha level was set to 0.05. When main effect or interaction effect were detected, Bonferroni adjustments were used for pairwise comparison, so that the alpha level was divided by the number of comparisons.

Results

Joint Stiffness

In both WRT and RWT, joint stiffness (K_{joint}) joint × step interaction effect was significant (p < .0001), and so a simple pairwise comparison was conducted (adjusted α = 0.0011). In WRT, K_{hip} was higher than K_{knee} at S1 (p = .0002), and K_{ankle} was higher than K_{knee} at S2 (p = .0004) and S1 (p = .001) (Table 5.1). Within K_{ankle}, S2 was higher than S-2 (p = .0004), S-1 (p = .0007) and S0 (p = .0001); within K_{hip}, S1 was higher than S-2 (p = .0002) and S-1 (p = .0001) (Table 5.1). In RWT, K_{knee} was lower than K_{ankle} (p < .0001, p = .0002) and K_{hip} (p = .0005, p < .0001) at S-2 and S-1, respectively (Table 5.1). Within K_{ankle}, S2 was lower than S-2 (p = .0009) and S-1 (p = .0006); S-1 was higher than S1 (p = .001) and S2 (p = .0003) within K_{hip} (Table 5.1).

Table 5.1. Joint stiffness (Nm/kg/deg) across WRT and RWT steps. Sample Mean (SD); n = 10.

Joint Stiffness (Nm/kg/deg)	**Steps**				
	S-2	**S-1**	**S0**	**S1**	**S2**
WRT					
Ankle	0.13 (0.05)[c]	0.12 (0.04)	0.16 (0.09)[c]	0.23 (0.11)[b]	0.24 (0.10)[b,c]
Knee	0.11 (0.05)	0.10 (0.03)	0.11 (0.06)	0.11 (0.05)[a,b]	0.11 (0.06)[b]
Hip	0.08 (0.03)[d]	0.08 (0.02)[d]	0.17 (0.09)	0.20 (0.06)[a,d]	0.24 (0.12)
RWT					
Ankle	0.21 (0.09)[e]	0.18 (0.07)[e]	0.19 (0.10)	0.15 (0.06)	0.13 (0.06)
Knee	0.12 (0.08)[e,f]	0.11 (0.07)[e,f]	0.12 (0.07)	0.12 (0.07)	0.09 (0.04)
Hip	0.21 (0.12)[f]	0.18 (0.07)[f,g]	0.12 (0.05)	0.11 (0.05)[g]	0.08 (0.05)[g]

a: Statistically significant differences between K_{hip} and K_{knee} at S1 in WRT, (p = .0002);
b: Differences between K_{ankle} and K_{knee} at S2 (p = .0004) and S1 (p = .001), respectively in WRT;
c: Differences between S2 and S-2 (p = .0004), S2 and S-1 (p = .0007), S2 and S0 (p = .0001) in WRT, respectively within K_{ankle};
d: Differences between S1 and S-2 (p = .0002), S1 and S-1 (p = .0001) in WRT, respectively within K_{hip};
e: Differences between K_{ankle} and K_{knee} at S-2 ($p < .0001$) and S-1 (p = .0002), respectively in RWT;
f: Differences between K_{hip} and K_{knee} at S-2 (p = .0005) and S-1 ($p < .0001$), respectively in RWT;
g: Differences between S-1 and S1 (p = .001), S-1 and S2 (p = .0003) in RWT, respectively within K_{hip}.

Joint Work

For W_{joint}^{+} and W_{joint}^{-} in stance and swing phase within WRT and RWT, joint × step interaction effects were all significant (p < .001), and so a simple pairwise comparison was conducted (adjusted α = 0.0011). In WRT, stance phase W_{ankle}^{+} was higher than W_{knee}^{+} at S-2 (p = .0003), S-1 (p = .0001) and S2 (p = .0003), W_{ankle}^{+} was also higher than W_{hip}^{+} at S1 (p = .0002) and S2 (p = .0001), W_{knee}^{+} was higher than W_{hip}^{+} at S1 (p = .0005) and S2 (p = .001) (Table 5.2). Within W_{hip}^{+} in stance, S-2 was higher than S1 (p < .0001) and S2 (p = .0003), S-1 was higher than S0 (p = .0006), S1 (p < .0001) and S2 (p < .0001) (Table 5.2). Stance phase W_{knee}^{-} was higher than W_{ankle}^{-} at S-1 (p < .0001), W_{knee}^{-} was also higher than W_{hip}^{-} at S1 (p = .0009) and S2 (p = .001) (Table 5.2). Within W_{ankle}^{-} in stance, S-2 was lower than S1 (p < .0001) and S2 (p < .0001), S-1 was lower than S1 (p < .0001) and S2 (p < .0001) (Table 5.2). In WRT swing phase, W_{hip}^{+} was higher than W_{ankle}^{+} and W_{knee}^{+} at all steps (between S-2 and S2) (p < .0001); within W_{hip}^{+}, S-2 was lower than S1 (p = .0004) (Table 5.2). Swing phase W_{knee}^{-} was higher than W_{ankle}^{-} and W_{hip}^{-} at all steps (p < .0001); among steps between S-1 and S1, W_{hip}^{-} was higher than W_{ankle}^{-} (p < .001); within W_{knee}^{-}, S-2 and S-1 were lower than all steps between S0 and S2, respectively (p < .0001) (Table 5.2).

In RWT, stance phase W_{ankle}^{+} was higher than W_{knee}^{+} and W_{hip}^{+} at all steps between S-2 and S0, respectively (p < .0011); W_{knee}^{+} was higher than W_{hip}^{+} at S-2 and S-1 (p < .0011) (Table 5.3). Within W_{ankle}^{+} in stance, S-2 was higher than S2 (p = .0005); within W_{knee}^{+} S-1 was higher than S1 (p = .001); within W_{hip}^{+} S-2 and S-1 were lower than steps between S1 and S2 (p < .0011) (Table 5.3). Stance phase W_{ankle}^{-} was lower than W^{-}_{knee} at S1 (p = .0006) and S2 (p < .0001); within W^{-}_{ankle}, S-2 and S-1 were higher than all steps between S0 and S2 (p < .0011) (Table 5.3). In RWT swing phase, many significant differences were detected however; only one comparison was insignificant at S1 (between W^{+}_{hip} and W^{+}_{knee}, p = .01). W^{+}_{hip} was higher than W^{+}_{ankle} and W^{+}_{knee} among all other steps between S-2 and S2 (p < .0011), $W^{+}_{ankle}$was higher than W^{+}_{knee} at S-1 (p = .0002). For W^{+}_{ankle} in swing, S-1 was higher than S2 (p = .0003) (Table 5.3). Swing phase W^{-}_{knee} was higher than W^{-}_{ankle} and W^{-}_{hip} at all steps, respectively (p < .0001); W^{-}_{hip} was also higher than W^{-}_{ankle} at all steps between S-1 and S2 (p < .0011); within W^{-}_{knee}, S2 was lower than S-2 (p = .0004) and S-1 (p < .0001), and S0 was lower than S-1 (p = .0003) (Table 5.3).

Table 5.2. Joint work (J/kg) across WRT steps. Sample Mean (SD); n = 10.

Joint Work (J/kg)	**Steps**				
	S-2	**S-1**	**S0**	**S1**	**S2**
Stance Phase Positive Work					
Ankle	0.40 (0.16)[a]	0.39 (0.13)[a]	0.55 (0.25)	0.63 (0.29)[b]	0.61 (0.23)[a,b]
Knee	0.21 (0.08)[a]	0.20 (0.08)[a]	0.37 (0.19)	0.31 (0.15)[c]	0.28 (0.14)[a,c]
Hip	0.20 (0.07)[d]	0.24 (0.08)[e]	0.09 (0.09)[e]	0.05 (0.05)[b,c,d,e]	0.06 (0.03)[b,c,d,e]
Stance Phase Negative work					
Ankle	0.10 (0.04)[h]	0.11 (0.06)[f,i]	0.28 (0.13)	0.36 (0.12)[h,i]	0.36 (0.08)[h,i]
Knee	0.30 (0.21)	0.24 (0.06)[f]	0.41 (0.32)	0.43 (0.18)[g]	0.41 (0.21)[g]
Hip	0.17 (0.12)	0.15 (0.08)	0.22 (0.18)	0.24 (0.19)[g]	0.19 (0.19)[g]
Swing Phase Positive Work					
Ankle[j]	0.01 (0.00)	0.01 (0.00)	0.01 (0.00)	0.01 (0.00)	0.01 (0.00)
Knee[j]	0.02 (0.02)	0.02 (0.01)	0.01 (0.01)	0.01 (0.01)	0.03 (0.04)
Hip[j]	0.14 (0.03)[k]	0.13 (0.06)	0.21 (0.07)	0.23 (0.07)[k]	0.26 (0.11)
Swing Phase Negative Work					
Ankle[l]	0.01 (0.03)	<0.01 (0.00)[m]	<0.01 (0.00)[m]	<0.01 (0.00)[m]	0.03 (0.10)
Knee[l]	0.23 (0.03)[n]	0.24 (0.04)[o]	0.31 (0.05)[n,o]	0.34 (0.06)[n,o]	0.35 (0.06)[n,o]
Hip[l]	0.03 (0.02)	0.02 (0.01)[m]	0.02 (0.01)[m]	0.02 (0.01)[m]	0.02 (0.01)

Note: < 0.01 indicates a negligible value; Joint negative work data were presented in absolute values.

a: Statistically significant differences between W^{+}_{ankle} and W^{+}_{knee} at S-2 (p = .0003), S-1 (p = .0001) and S2 (p = .0003), respectively;

b: Differences between W^{+}_{ankle} and W^{+}_{hip} at S1 (p = .0002) and S2 (p = .0001) at stance, respectively;

c: Differences between W^{+}_{knee} and W^{+}_{hip} at S1 (p = .0005) and S2 (p = .001) at stance, respectively;

d: Differences between S-2 and S1 (p < .0001), S-2 and S2 (p = .0003) at stance, respectively within W^{+}_{hip};

e: Differences between S-1 and S0 (p = .0006), S-1 and S1 (p < .0001), S-1 and S2 (p < .0001) at stance, respectively within W_{hip}^{+};

f: Differences between W_{ankle}^{-} and W_{knee}^{-} at S-1 at stance, ($p < .0001$);

g: Differences between W_{knee}^{-} and W_{hip}^{-} at S1 ($p = .0009$) and S2 ($p = .001$) at stance, respectively;

h: Differences between S-2 and S1 ($p < .0001$), S-2 and S2 ($p < .0001$) at stance, respectively within W_{ankle}^{-};

i: Differences between S-1 and S1 ($p < .0001$), S-1 and S2 ($p < .0001$) at stance, respectively within W_{ankle}^{-};

j: Differences between W_{hip}^{+} and W_{ankle}^{+}, and W_{hip}^{+} and W_{knee}^{+} at all steps at swing, ($p < .0001$);

k: Differences between S-2 and S1 at swing within W_{hip}^{+}, ($p = .0004$);

l: Differences between W_{knee}^{-} and W_{ankle}^{-}, and W_{knee}^{-} and W_{hip}^{-} at swing at all steps, ($p < .0001$);

m: Differences between W_{ankle}^{-} and W_{hip}^{-} at swing, respectively at all steps between S-1 and S1, ($p < .001$);

n: Differences between S-2 and all steps between S0 and S2 at swing, respectively within W_{knee}^{-}, ($p < .0001$);

o: Differences between S-1 and all steps between S0 and S2 at swing, respectively within W_{knee}^{-}, ($p < .0001$).

Table 5.3. Joint work (J/kg) across RWT steps. Sample Mean (SD); n = 10.

Joint Work (J/kg)	Steps				
	S-2	**S-1**	**S0**	**S1**	**S2**
Stance Phase Positive Work					
Ankle	0.67 (0.31)[a,b,d]	0.66 (0.33)[a,b]	0.50 (0.21)[a,b]	0.48 (0.34)	0.37 (0.20)[d]
Knee	0.33 (0.16)[a,c]	0.32 (0.15)[a,c,e]	0.21 (0.14)[a]	0.23 (0.17)[e]	0.19 (0.11)
Hip	0.06 (0.04)[b,c,f]	0.07 (0.05)[b,c,g]	0.16 (0.08)[b]	0.20 (0.06)[f,g]	0.18 (0.04)[f,g]
Stance Phase Negative work					
Ankle	0.41 (0.19)[i]	0.38 (0.18)[j]	0.17 (0.12)[i,j]	0.12 (0.08)[h,i,j]	0.10 (0.05)[h,i,j]
Knee	0.41 (0.21)	0.45 (0.24)	0.42 (0.22)	0.28 (0.10)[h]	0.26 (0.06)[h]
Hip	0.25 (0.22)	0.26 (0.18)	0.23 (0.18)	0.22 (0.15)	0.18 (0.11)
Swing Phase Positive Work					
Ankle[k]	0.01 (0.00)	0.01 (0.00)[l]	0.01 (0.00)	0.01 (0.01)	0.01 (0.00)
Knee[k]	0.01 (0.01)	<0.01 (0.00)[l]	0.01 (0.01)	0.02 (0.04)	0.01 (0.01)
Hip[k]	0.22 (0.09)	0.22 (0.09)	0.13 (0.04)	0.12 (0.06)	0.12 (0.03)
Swing Phase Negative Work					
Ankle[m]	<0.01 (0.00)	<0.01 (0.00)[n]	<0.01 (0.00)[n]	<0.01 (0.00)[n]	<0.01 (0.00)[n]
Knee[m]	0.33 (0.09)[o]	0.34 (0.08)[o,p]	0.23 (0.05)[p]	0.21 (0.08)	0.21 (0.04)[o]
Hip[m]	0.01 (0.01)	0.02 (0.01)[n]	0.02 (0.01)[n]	0.02 (0.01)[n]	0.02 (0.01)[n]

Note: < 0.01 indicates a negligible value; Joint negative work data were presented in absolute values.

a: Statistically significant differences between W^{+}_{ankle} and W^{+}_{knee} at all steps between S-2 and S0 at stance, ($p < .0011$);

b: Differences between W^{+}_{ankle} and W^{+}_{hip} at all steps between S-2 and S0 at stance, ($p < .0011$);

c: Differences between W^{+}_{knee} and W^{+}_{hip} at S-2 and S-1 at stance, respectively ($p < .0011$);

d: Differences between S-2 and S2 at stance within W^{+}_{ankle}, ($p = .0005$);

e: Differences between S-1 and S1 at stance within W^{+}_{knee}, ($p = .001$);

f: Differences between S-2 and steps between S1 and S2 at stance, respectively within W^{+}_{hip} (p < .0011);
g: Differences between S-1 and steps between S1 and S2 at stance, respectively within W^{+}_{hip} (p < .0011);
h: Differences between W^{-}_{ankle} and W^{-}_{knee} at S1 (p = .0006) and S2 (p < .0001) at stance, respectively;
i: Differences between S-2 and steps between S0 and S2 at stance within W^{-}_{ankle}, respectively (p < .0011);
j: Differences between S-1 and steps between S0 and S2 at stance within W^{-}_{ankle}, respectively (p < .0011);
k: Differences between W^{+}_{hip} and W^{+}_{ankle}, W^{+}_{hip} and W^{+}_{knee} at all steps at swing, except for W^{+}_{knee} at S1, (p < .0011);
l: Differences between W^{+}_{ankle} and W^{+}_{knee} at S-1 at stance, (p = .0002);
m: Differences between W^{-}_{knee} and W^{-}_{ankle}, W^{-}_{knee} and W^{-}_{hip} at all steps at swing, respectively (p < .0001);
n: Differences between W^{-}_{ankle} and W^{-}_{hip} at steps between S-1 and S2 at swing, respectively (p < .0011);
o: Differences between S2 and S-2 (p = .0004), S2 and S-1 (p < .0001) at swing within W^{-}_{knee}, respectively;
p: Differences between S-1 and S0 at swing within W^{-}_{knee}, (p = .0003).

Table 5.4. Joint stance phase extensor moment angular impulse (Nm·s/kg) and total support torque (Nm·s/kg) across WRT and RWT steps. Sample Mean (SD); n = 10.

Impulse (Nm·s/kg)	Steps				
	S-2	S-1	S0	S1	S2
WRT					
Ankle[a]	0.40 (0.12)[b]	0.36 (0.07)[b]	0.38 (0.12)	0.41 (0.15)	0.40 (0.11)
Knee	0.16 (0.12)[b,d]	0.14 (0.08)[b,e]	0.29 (0.19)	0.30 (0.14)[c,d,e]	0.26 (0.15)[d]
Hip[a]	0.12 (0.03)[f]	0.12 (0.02)[g]	0.10 (0.04)[h]	0.05 (0.03)[c,f,g,h]	0.07 (0.02)[f,g]
Total	0.68 (0.23)	0.62 (0.12)	0.77 (0.28)	0.76 (0.28)	0.73 (0.24)
RWT					
Ankle[i]	0.46 (0.22)	0.44 (0.21)	0.41 (0.19)	0.44 (0.24)	0.37 (0.15)
Knee	0.30 (0.15)[j,k]	0.32 (0.15)[j,k]	0.25 (0.13)	0.18 (0.10)[k]	0.16 (0.08)
Hip[i]	0.06 (0.03)[j,l]	0.06 (0.04)[j,m]	0.08 (0.05)	0.12 (0.04)[l]	0.11 (0.04)[l,m]
Total	0.82 (0.36)	0.82 (0.36)	0.73 (0.33)	0.75 (0.37)	0.65 (0.24)

a: Statistically significant differences between I_{ankle} and I_{hip} at all steps in WRT, respectively (p < .0011);

b: Differences between I_{ankle} and I_{knee} at S-2 (p < .0001) and S-1 (p < .0001) in WRT, respectively;

c: Differences between I_{knee} and I_{hip} at S1 in WRT, (p = .0005);

d: Differences between S-2 and S1 (p < .0001), S-2 and S2 (p = .0006) in WRT, respectively within I_{knee};

e: Differences between S-1 and S1 in WRT within I_{knee}, (p = .0008);

f: Differences between S-2 and S1, S-2 and S2 in WRT, respectively within I_{hip}, (p < .001);

g: Differences between S-1 and S1, S-1 and S2 in WRT, respectively within I_{hip}, (p < .001);

h: Differences between S0 and S1 in WRT within I_{hip}, (p = .0005);

i: Differences between I_{ankle} and I_{hip} at all steps in RWT, (p < .001);

j: Differences between I_{knee} and I_{hip} at S-2 (p = .0005) and S-1 (p = .0004) in RWT, respectively;

k: Differences between S1 and S-2 (p = .0005), S1 and S-1 (p = .0001) in RWT, respectively within I_{knee};

l: Differences between S-2 and S1 (p = .0003), S-2 and S2 (p < .0001) in RWT, respectively within I_{hip};

m: Differences between S-1 and S2 in RWT within I_{hip}, (p = .0002).

Joint Angular Impulse and Total Support Torque

In WRT stance phase, I_{ankle} was higher than I_{hip} at all steps, respectively ($p <$ .0011), I_{ankle} was higher than I_{knee} at S-2 ($p < .0001$) and S-1 ($p < .0001$), and I_{knee} was higher than I_{hip} at S1 ($p = .0005$); within I_{knee}, S-2 was lower than S1 ($p < .0001$) and S2 ($p = .0006$), and S-1 was lower than S1 ($p = .0008$); S-2 and S-1 was higher than all steps between S1 and S2 ($p < .001$), and S0 was higher than S1 ($p = .0005$) within I_{hip} (Table 5.4). In RWT stance phase, I_{ankle} was higher than I_{hip} at all steps ($p < .001$), and I_{knee} was higher than I_{hip} at S-2 ($p = .0005$) and S-1 ($p = .0004$); within I_{knee}, S1 was lower than S-2 ($p = .0005$) and S-1 ($p = .0001$); within I_{hip}, S-2 was lower than S1 ($p = .0003$) and S2 ($p < .0001$), S-1 was lower than S2 ($p = .0002$) (Table 5.4). No significant difference was found for I_{total} between different steps within WRT and RWT, respectively.

Joint Peak Power

For joint peak power comparison between steps, when step main effect was detected, follow up pairwise comparison was conducted (adjusted $\alpha = 0.005$). In WRT, ankle joint stance peak dorsiflexion power at S2 was higher than S-2 ($p < .0001$) and S-1 ($p < .0001$), and S1 was higher than S-2 ($p = .0009$) and S-1 ($p = .0003$) (Table 5.5). For stance phase peak knee flexion power, S1 was higher than S-2 ($p = .001$) and S-1 ($p =$ .003) (Table 5.5). Within stance phase peak hip extension power, S-1 was higher than S1 ($p = .002$) (Table 5.5). In RWT, stance phase peak ankle dorsiflexion power at S-2 was higher than S0 ($p = .002$), S1 ($p < .001$) and S2 ($p < .001$), and S-1 was also higher than S1 ($p < .0001$) and S2 ($p = .0005$) (Table 5.6). Swing phase peak ankle dorsiflexion power at S-2 was lower than S1 (p = .001) and S2 (p = .003), and S-1 was lower than S1 (p = .002) (Table 5.6).

Table 5.5. Joint peak power (W/kg) across WRT steps. Sample Mean (SD); n = 10.

Joint Power (W/kg)	**Steps**				
	S-2	**S-1**	**S0**	**S1**	**S2**
Stance Phase Peak Flexion Power					
Ankle	-0.99 (0.27)[a,b]	-1.18 (0.60)[a,b]	-3.65 (2.04)	-5.56 (2.19)[b]	-5.71 (1.26)[a]
Knee	-3.43 (2.02)[c]	-2.86 (0.99)[c]	-6.22 (5.25)	-7.39 (3.42)[c]	-7.60 (3.77)
Hip	-1.37 (0.74)	-1.36 (0.54)	-2.45 (2.09)	-2.82 (2.07)	-2.43 (2.19)
Stance Phase Peak Extension Power					
Ankle	5.68 (2.68)	4.95 (2.23)	6.38 (3.54)	8.06 (4.51)	8.09 (3.68)
Knee	3.61 (1.08)	3.14 (0.91)	4.90 (3.08)	4.62 (2.47)	4.26 (2.09)
Hip	2.09 (0.83)	2.23 (0.53)[d]	1.26 (0.75)	1.14 (0.62)[d]	1.37 (0.57)
Swing Phase Peak Flexion Power					
Ankle	-0.11 (0.06)	-0.07 (0.05)	-0.08 (0.03)	-0.07 (0.03)	-0.08 (0.03)
Knee	-2.29 (0.40)	-2.39 (0.40)	-2.45 (0.47)	-2.73 (0.42)	-2.92 (0.45)
Hip	-0.43 (0.53)	-0.37 (0.14)	-0.34 (0.19)	-0.28 (0.12)	-0.37 (0.17)
Swing Phase Peak Extension Power					
Ankle	0.11 (0.03)	0.08 (0.03)	0.09 (0.03)	0.09 (0.03)	0.09 (0.03)
Knee	0.81 (0.43)	0.89 (0.33)	0.79 (0.31)	0.85 (0.45)	0.97 (0.54)
Hip	1.96 (0.60)	1.38 (0.51)	1.28 (0.49)	1.22 (0.39)	1.49 (1.15)

Note: We defined extension direction as positive while flexion as negative among all three joints net joint power curve in this study.

a: Statistically significant differences between S2 and S-2 ($p < .0001$), S2 and S-1 ($p < .0001$), respectively within ankle joint stance phase peak dorsiflexion power;

b: Differences between S1 and S-2 ($p = .0009$), S1 and S-1 ($p = .0003$), respectively within ankle joint stance phase peak dorsiflexion power;

c: Differences between S1 and S-2 ($p = .001$), S1 and S-1 ($p = .003$), respectively within knee joint stance phase peak flexion power;

d: Differences between S-1 and S1 within hip stance phase peak extension power, (p = .002).

Table 5.6. Joint peak power (W/kg) across RWT steps. Sample Mean (SD); n = 10.

Joint Power (W/kg)	**Steps**				
	S-2	**S-1**	**S0**	**S1**	**S2**
Stance Phase Peak Flexion Power					
Ankle	-5.76 (2.13)[a]	-5.11 (2.14)[b]	-1.98 (1.16)[a]	-1.47 (1.31)[a,b]	-1.02 (0.55)[a,b]
Knee	-7.00 (3.62)	-6.92 (3.29)	-4.62 (2.14)	-3.41 (1.29)	-2.81 (0.57)
Hip	-2.81 (2.69)	-2.59 (1.82)	-1.88 (1.25)	-1.61 (0.63)	-1.41 (0.37)
Stance Phase Peak Extension Power					
Ankle	7.93 (4.15)	7.17 (3.75)	4.16 (1.84)	5.67 (3.32)	4.64 (1.28)
Knee	4.52 (2.98)	4.35 (2.31)	2.75 (1.48)	2.96 (1.46)	2.60 (0.93)
Hip	1.74 (1.25)	1.51 (1.09)	2.29 (0.86)	2.27 (0.77)	1.85 (0.42)
Swing Phase Peak Flexion Power					
Ankle	-0.06 (0.04)[c]	-0.06 (0.04)[d]	-0.06 (0.04)	-0.10 (0.05)[c,d]	-0.10 (0.05)[c]
Knee	-2.69 (0.58)	-2.81 (0.54)	-2.38 (0.41)	-2.17 (0.63)	-2.15 (0.28)
Hip	-0.25 (0.12)	-0.30 (0.12)	-0.32 (0.15)	-0.27 (0.14)	-0.24 (0.14)
Swing Phase Peak Extension Power					
Ankle	0.09 (0.02)	0.10 (0.02)	0.09 (0.03)	0.12 (0.11)	0.08 (0.02)
Knee	0.56 (0.36)	0.36 (0.21)	0.37 (0.32)	0.75 (0.77)	0.52 (0.31)
Hip	1.19 (0.43)	1.26 (0.40)	1.43 (0.44)	1.34 (0.49)	1.49 (0.23)

Note: We defined extension direction as positive while flexion as negative among all three joints net joint power curve in this study.

a: Statistically significant differences between S-2 and S0 (p = .002), S-2 and S1 (p < .001), S-2 and S2 (p < .001), respectively within ankle joint stance phase peak dorsiflexion power;

b: Differences between S-1 and S1 (p < .0001), S-1 and S2 (p = .0005), respectively within ankle joint stance phase peak dorsiflexion power;

c: Differences between S-2 and S1 (p = .001), S-2 and S2 (p = .003), respectively within ankle joint swing phase peak dorsiflexion power;

d: Differences between S-1 and S1 within ankle joint swing phase peak dorsiflexion power, (p = .002).

Curve Patterns

In WRT, the magnitude of flexion and extension power for the ankle, knee and hip joints tended to increase from S0 to S2 (Fig. 5.1). However, in RWT, the magnitude of three joints power decreased from S0 to S2 (Fig. 5.2). In WRT, net joint power characteristics in all three joints at transition step S0 tended to be similar to running gait patterns (Fig. 5.1). At S0 for RWT process, knee joint power characteristics tended to be similar to walking gait patterns, while ankle and hip joint power patterns tended to be a combination of both walking and running (Fig. 5.2). Similar trends were also found in the ankle joint angle-moment curves for WRT and RWT steps (Fig. 5.3).

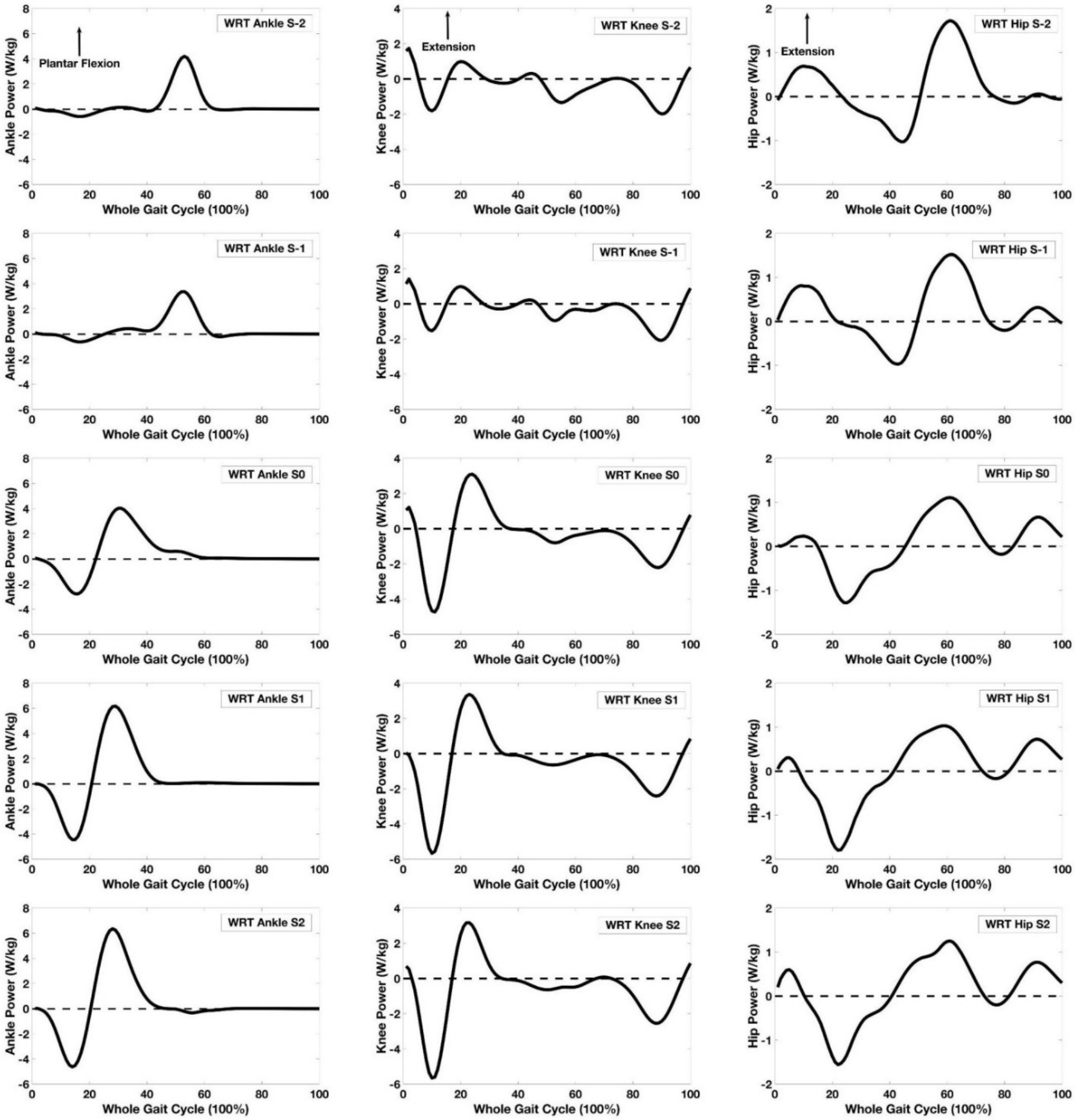

Fig 5.1. Group average (n = 10) ankle, knee and hip joint power curves in WRT steps, time normalize to whole gait cycle.

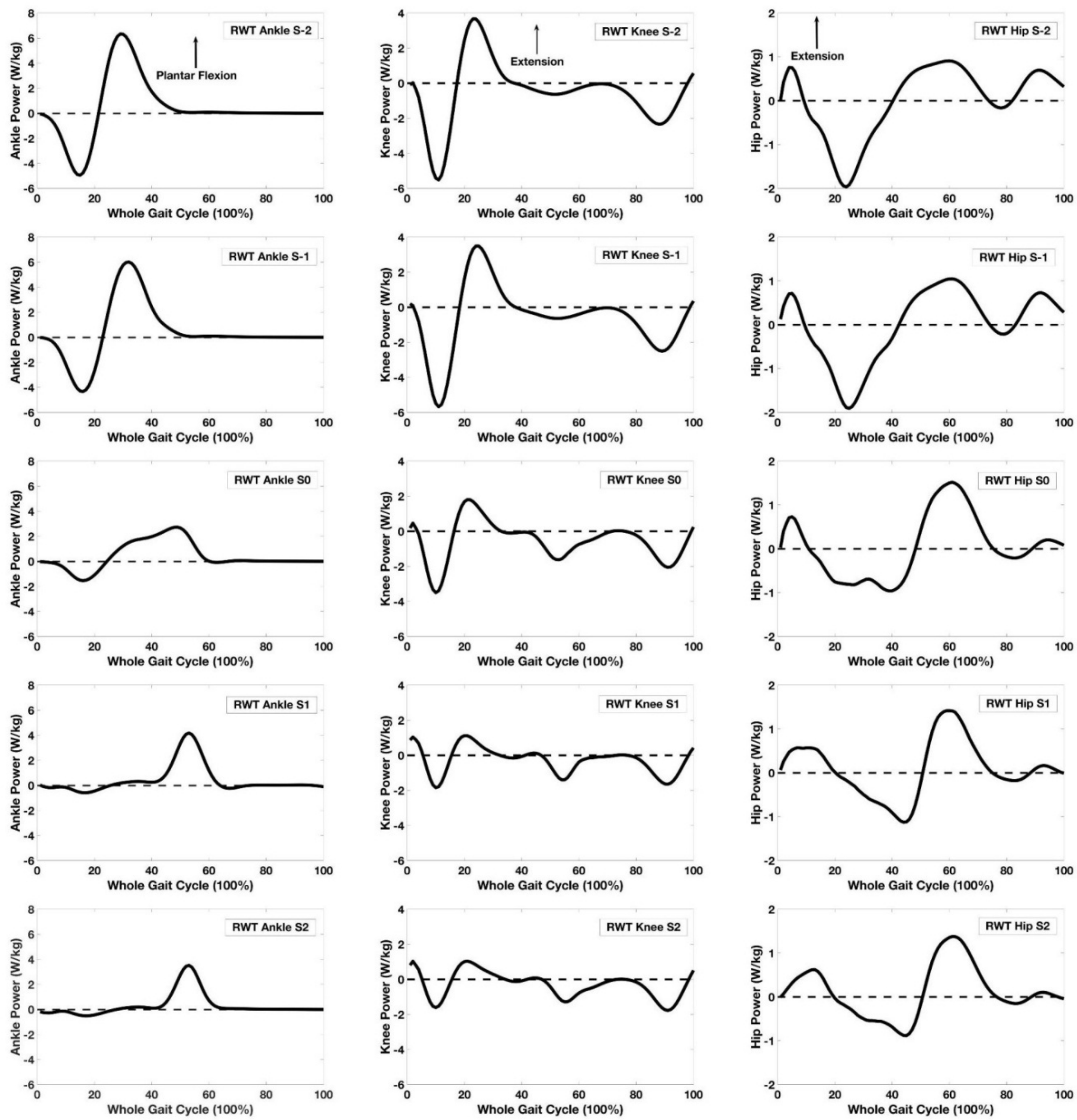

Fig 5.2. Group average (n = 10) ankle, knee and hip joint power curves in RWT steps, time normalize to whole gait cycle.

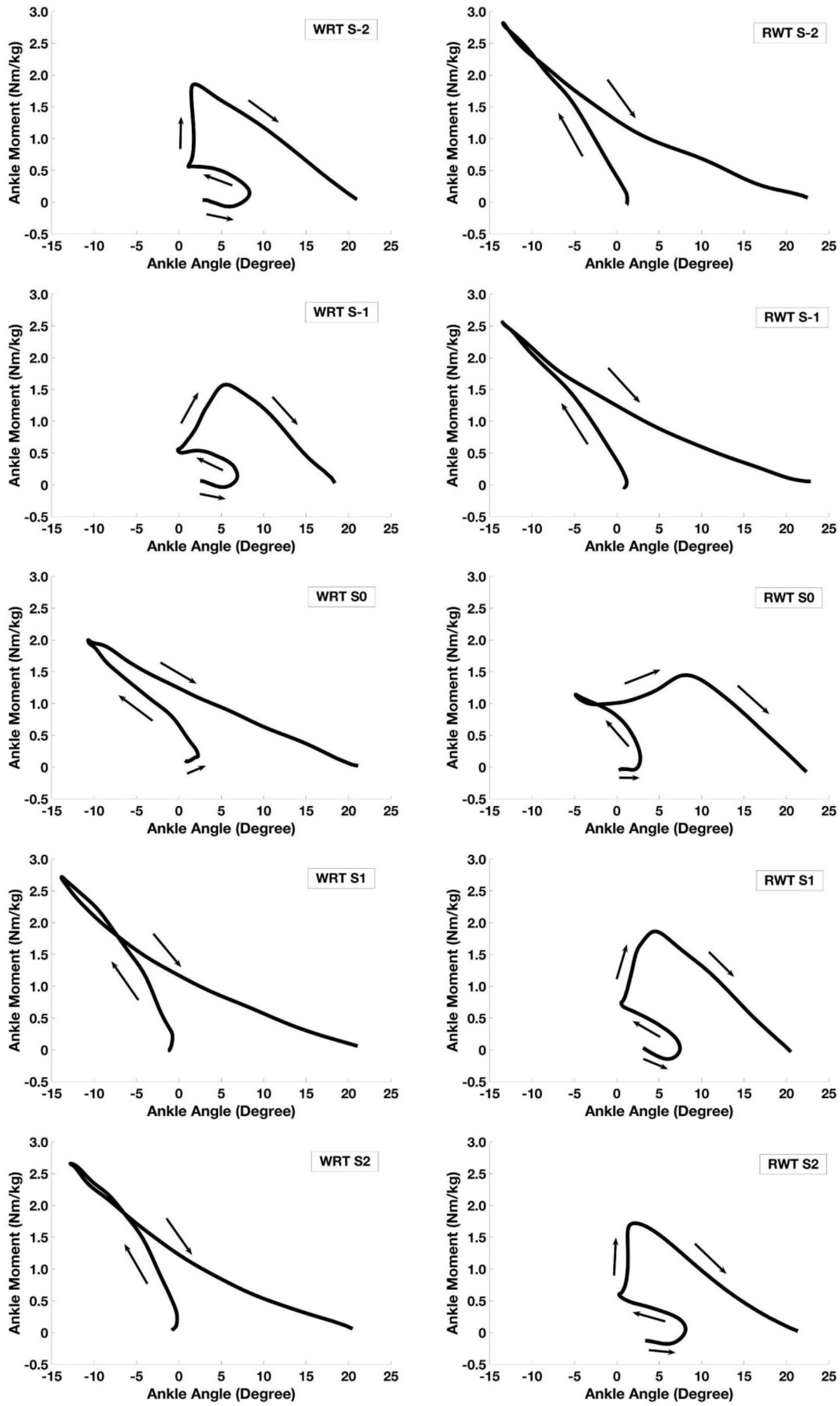

Fig 5.3. Group average (n = 10) ankle joint stance phase angle-moment curves in WRT steps (left) and RWT steps (right).

Discussion

This study aimed to investigate lower extremity joint level kinetic characteristics across steps in the WRT and RWT processes. Specifically, the goal was to better understand lower extremity joint stance phase dynamic loading, mechanical energy absorption and generation, as well as functional roles between three joints in both stance and swing phase from pre-transition to post-transition process. The initial hypothesis that K_{joint} would increase in WRT and decrease in RWT was partially supported. Specifically, K_{knee} tended to remain unchanged across WRT and RWT steps, K_{ankle} and K_{hip} tended to increase from S0 to S2 in WRT, and decrease from S0 to S2 in RWT (Table 5.1). This indicates that the transition between walking and running has more influence on dynamic loading and response via K_{ankle} and K_{hip}.

The initial hypothesis that W_{joint}^{+}, W_{joint}^{-} and I_{joint} would increase in WRT and decrease in RWT was also partially supported. In WRT stance phase, W_{ankle}^{+} and W_{knee}^{+} were 34% and 60% higher at S0 compared with at S-1, respectively; while for W_{hip}^{+} there was a significant decrease (p = .0006, 91% decrease) at S0 compared with S-1 (Table 5.2). Furthermore, W_{ankle}^{+} and W_{knee}^{+} were higher compared with W_{hip}^{+} within steps between S1 and S2, indicating that when switching from walking to running gait, the ankle and knee joint played more dominant roles in stance phase energy generation from the transition step (S0) and the following steps (S1, S2). This agreed with previous reports where the ankle and knee positive mechanical work and power were regarded as important factors which drive the WRT process (V. Segers et al., 2013). Decreasing W_{hip}^{+} while increasing of W_{ankle}^{+} within steps from S0 to S2 compared with previous steps before the transition may indicate that during the WRT process, especially from the transition step, stance phase energy generation tended to transfer from proximal to distal. The W_{ankle}^{-} during the pre-transition steps (S-2, S-1) was significantly lower than during the post-transition steps (S1, S2), and W_{ankle}^{-} at S0 was 87% higher than at S-1 (Table 5.2). This indicates that the ankle joint absorbs more energy when transitioning from walking to running gait pattern. The knee joint played an important role in energy absorption of WRT stance phase, with W_{knee}^{-} at S0 calculated to be 52% higher than at S-1 (Table 5.2).

In RWT stance phase, the ankle and knee joint played more dominant roles in energy generation within the pre-transition steps (S-2, S-1) (Table 5.3). The W_{ankle}^{+} was significantly higher than W_{knee}^{+} and W_{hip}^{+} within all steps between S-2 and S0 (Table 5.3). Additionally, W_{ankle}^{+} and W_{knee}^{+} at S0 was 28% and 42% lower than at S-1, respectively; while W_{hip}^{+} at S0 was 78% higher than at S-1 (Table 5.3). This indicates that in RWT stance phase, energy generation tended to transfer from distal to proximal, and that W_{ankle}^{+} and W_{knee}^{+} decrease while W_{hip}^{+} increases during and after a transition to a walking pattern. The energy generation transfer phenomenon among lower extremity joints at the transition step (S0) in both WRT and RWT may be due to the idea that lower extremity distal joints have a higher stance phase mechanical work generation efficiency, or need less effort when running at speeds right above subjects' preferred transition speed (PTS) (Farris & Sawicki, 2012; Pires et al., 2014), and vice versa for walking at speeds below PTS. The stance phase energy generation transfer mechanism was sensitive at transition step (S0) for both WRT and RWT. The lower extremity energy generation transfer mechanism can be attributed to the combined choice of gait transition trigger mechanisms: optimization of mechanical work efficiency, as well as minimizing of musculoskeletal system effort ("effort-load hypothesis") at transition step between walking and running (Farris & Sawicki, 2012; Pires et al., 2014). Similar to WRT, W_{ankle}^{-} was lower at steps after the transition from running to walking. And W_{ankle}^{-} at S0 was 76% lower than at S-1 (Table 5.3). This may indicate stance phase W_{ankle}^{-} was sensitive to gait pattern changes between walking and running. The knee joint played an important function in RWT stance phase energy absorption. In swing phase in both WRT and RWT, hip and knee joints were playing dominant roles in energy generation and absorption, respectively. Swing phase W_{hip}^{+} and W_{knee}^{-} tended to increase in the WRT process and decrease in the RWT process (Table 5.2, Table 5.3).

The I_{ankle} played an important role in both WRT and RWT processes, especially during the walking steps in both transition cases, compared with knee and hip joint. In WRT and RWT, I_{ankle} tended to be unchanged across steps (Table 5.4), while I_{knee} and I_{hip} were more affected by gait pattern changes between walking and running. In WRT, I_{knee} at S0 was 70% higher than at S-1, indicating knee joint extensor moment plays an important role in WRT. Further, I_{knee} and I_{hip} tended to have a reverse trend when switching gait patterns in both WRT and RWT process. Specifically, I_{knee} tended to increase from S0 to S2, while I_{hip} tended to decrease from S0 to S2 during the WRT process, and vice versa for the RWT process (Table 5.4).

In both WRT and RWT, changing of locomotion patterns had influence on stance phase ankle joint peak dorsiflexion power characteristics. Specifically, ankle peak dorsiflexion power increased in the WRT process, and decreased in RWT (Table 5.5, Table 5.6). This agreed with stance phase W_{ankle}^{-} characteristics in both transitions.

In this study, we investigated stance phase sagittal plane ankle joint angle-moment relationship among steps in both WRT and RWT (Fig. 5.3). Similar to previous studies, the stance phase ankle angle-moment relationship was a clockwise hysteresis curve (Crenna & Frigo, 2011; Gabriel et al., 2008; Kuitunen et al., 2002). Walking steps had three different phases: initial contact, ascending and descending phase. In this gait transition study, the ascending phase had two sub-phases: loading ascending and yielding ascending phase. Running steps had two phases: ascending and descending phase. The ascending phase can be regarded as the dynamic loading period, while the descending phase can be treated as the energy generation period for both walking and running (Crenna & Frigo, 2011).

At steps S-2 and S-1 in WRT, after initial contact, the ankle would first dorsiflex to a nearly neutral position (loading ascending phase). Then the ankle tended to remain "locked" in late middle stance period (yielding ascending phase) at S-2, as the plantarflexion moment quickly increased in this phase. As the plantar flexor moment then decreased, the ankle joint released the energy in the descending phase (push-off period). The yielding ascending pattern was different at S-1, with the ankle joint already starting a plantar flexion movement when the plantar flexor moment increased during the mid-stance period. This indicates that the ankle joint started the generation of energy earlier compared with the previous step (S-2), before the push-off period during late stance. The reason for this pattern may be attributed to the constant acceleration of speed experienced in the WRT protocol required the ankle joint to generate the energy at an earlier period, compared with a steady state fast walking speed condition. At transition step S0, the curve pattern abruptly changed, to be more similar with a running gait pattern. The initial contact period was shortened, and then the ankle would dorsiflex over the neutral position in the ascending phase, indicating a higher amount of energy absorption in this period compared with previous steps. Compared with steps at S1 and S2, a relatively wide open area between the ascending and descending phase of the curve at S0 showed that more energy generation was needed, compared to the amount of energy absorbed in the previous ascending phase period. Calculation of stance phase W_{ankle}^{+} and W_{ankle}^{-} ratio ($W_{ankle}^{+}/W_{ankle}^{-}$) from S0 to S2 provided further evidence: the ratio was 1.93 at S0, 1.74 at S1 and 1.69 at S2 (Table 5.2). At step S1 and S2 in WRT, both curves exhibited a typical running gait pattern with the ankle dorsiflexing more than during the transition step.

In the RWT process, both curves were identical at S-2 and S-1. The slope of both the ascending and descending phase curve tended to decrease at S-1 compared to S-2. Then the curve changed into a four-phases pattern at S0. After brief initial contact, the ankle dorsiflexed to a smaller angle in the ascending phase, compared with previous running steps. In the yielding ascending phase, the ankle began a plantar flexion movement with a slight increase of plantar flexor moment. This may be attributed to the constantly decelerating speed, along with an increase in stance phase time. This phase required less energy absorption and a longer stance time allows the ankle to generate energy. Compared with a typical walking condition, the energy generation period was still earlier at S0. Additionally, the ankle power curve at S0 showed a similar length of walking stance time, as well as early stance dorsiflexion power pattern. All these patterns indicated that the angle-moment curve at S0 was a combination of both gait patterns (Fig. 5.2, Fig. 5.3). At steps S1 and S2, the curves patterns were very similar to a high-speed walking gait pattern and were similar to S-2 in WRT. The transition step (S0) ankle angle-moment curve pattern was close to the running gait pattern in WRT; however, in the RWT process, the curve pattern appeared to be a combination of both walking and running conditions at the transition step. This indicates that RWT would take a longer time for subjects to adjust and modulate their motor response compared with the WRT process. The preferred transition speed (PTS) in WRT (2.06 ± 0.09 m/s) and RWT (1.97 ± 0.10 m/s) further indicates that it took longer for the subjects to finish the RWT process.

Limitations

One limitation of this study is that only one representative acceleration and deceleration value was used for WRT and RWT. Different acceleration and deceleration magnitudes would likely affect gait transition speed and transition step gait patterns (L. Li, 2000). Another limitation is that treadmill walking and running, using controlled locomotion speeds and treadmill acceleration and deceleration may be different from the naturally occurring patterns of over-ground gait transitions.

Future Work

Future studies should investigate gait transition patterns between different age groups, and examine the effect of a wider range of acceleration and deceleration magnitudes on the gait transition patterns. Additionally, it would be interesting to investigate whole body center of mass dynamics in both transition processes.

Conclusion

Both WRT and RWT have significant effects on ankle and hip joint stiffness characteristics for the transition and following steps. Stance phase energy generation tended to transfer from proximal to distal during WRT, while generation of energy transfer from distal to proximal during the RWT process. The stance phase mechanical energy generation transfer mechanism was sensitive at the transition step (S0) for both WRT and RWT. The stance phase lower extremity energy generation transfer mechanism may be the combined results of different gait transition trigger mechanisms. Ankle joint stance phase plantar flexor moment angular impulse played an important role in both WRT and RWT processes, while knee and hip joint extensor moment angular impulse tended to be influenced by gait pattern changes between walking and running. Joint power curve patterns at transition step were similar to target locomotion patterns in WRT. The transition step stance phase sagittal plane ankle joint angle-moment curve pattern was close to a running gait pattern in WRT, while the curve pattern appeared to be a combination between both walking and running condition at the transition step in RWT. These findings suggest that gait transition between walking and running affects lower extremity joint kinetic patterns.

Bridge

Chapter V demonstrated that gait transition between walking and running has significant effects on lower extremity joint kinetic patterns. While Chapter III-V mainly focused on lower extremity system in response to gait patterns and speed changes, little is known about the connection between whole body COM system and lower extremity system. Chapter VI incorporates the findings from Chapter III-V and delves deeper to investigate potential connection between the whole body COM and the lower extremity system. This could provide a broader perspective about locomotion dynamic patterns and potential gait efficiency optimization.

CHAPTER VI

LEG SPRING STIFFNESS AND WHOLE BODY CENTER OF MASS MECHANICAL WORK AND POWER IN RUNNING

This Chapter is currently unpublished. Li Jin designed this study, collected the data and analyzed it. Michael E. Hahn provided mentorship activities, including assistance with study design, general oversight of the project, and editing and finalizing of the journal manuscript.

Introduction

Running is a popular locomotion activity in human beings. It has a unique gait pattern compared with walking: single leg support in stance phase, followed by a flight phase within the gait cycle. Within running stance phase, the lower extremity is relatively compliant compared with walking (C T Farley & Ferris, 1998). In the first half of stance phase, lower extremity joints go through flexion movement; and extension movement during the second half of stance phase (C T Farley & Ferris, 1998). These motions suggest that in response to external moment and force, the lower extremity musculoskeletal system acts like a spring, absorbing energy in first half of stance and returning a portion of elastic energy in second half of stance (G. A. Cavagna et al., 1977, 1964; Claire T Farley & Gonzalez, 1996). This results in a unique pattern of motion for the whole body center of mass (COM) as well: the COM position reaches its minimum height at mid-stance phase and the COM movement trajectory is similar as a “bouncing ball” (G. A. Cavagna et al., 1964; C T Farley & Ferris, 1998) across stance phase. In the stance phase of running, the whole lower extremity system can be regarded as a “leg spring” due to its compliant behavior, and if the COM is regarded as a point mass, the system can be viewed as a “bouncing ball” attached to a “leg spring” (McGowan et al., 2012). Using this analogy, a simplified spring-mass model has been proposed and widely used in the analysis of human running gait (Brughelli & Cronin, 2008; C T Farley & Ferris, 1998; C T Farley et al., 1993; Ferris et al., 1998; McGowan et al., 2012; McMahon & Cheng, 1990).

The deformation and stretch characteristics (loading and unloading) of the “leg spring” system under external moment and force loading in running stance phase can be regarded as the stiffness pattern of lower extremity musculoskeletal system. Vertical stiffness (K_{vert}), leg stiffness (K_{leg}) and joint stiffness (K_{joint}) are three variables which can be directly calculated from running activities (Brughelli & Cronin, 2008). Moreover, K_{vert} and K_{leg} can be calculated via the spring-mass model mentioned previously. K_{vert} reflects COM vertical movement and oscillation characteristics in stance phase (Brughelli & Cronin, 2008; G. Cavagna et al., 1988; McMahon et al., 1987) and K_{vert} has been reported to increase with running speeds (Brughelli & Cronin, 2008; G. A. Cavagna, 2005; He et al., 1991; Morin et al., 2005). This may be attributed to an increase in peak vertical ground reaction force (GRF) while COM displacement decreases when running speeds increase (Brughelli & Cronin, 2008). For K_{leg}, this phenomenon reflects a length change link between the foot and the COM during ground contact (Brughelli & Cronin, 2008; C T Farley et al., 1993; McMahon & Cheng, 1990) and K_{leg} appears to remain relatively unchanged when running speeds increase (Biewener, 1989; Brughelli & Cronin, 2008; C T Farley et al., 1993; He et al., 1991; McMahon & Cheng, 1990; Morin et al., 2005, 2006). The behavior of K_{joint} reflects joint level intersegmental displacement as a function of joint moment loading (Crenna & Frigo, 2011; Davis & DeLuca, 1996; Gabriel et al., 2008; Jin & Hahn, 2018). It has been reported that K_{ankle} remains relatively unchanged when running speeds increase (Arampatzis et al., 1999; Kuitunen et al., 2002), while K_{knee} increases with running speeds (Arampatzis et al., 1999; Kuitunen et al., 2002). Most previous studies have investigated K_{vert} and K_{leg} together without incorporating K_{joint} into the same analysis. Thus, it remains that little is known about the relationship between K_{vert} and K_{joint}, or K_{leg} and K_{joint} across a range of running speeds, as well as whether K_{vert} and K_{leg} can be predicted from K_{joint} in different running speeds. From the previous findings reviewed in this section, it can be surmised that K_{vert} and K_{leg} patterns may emerge from local joint level elasticity (or stiffness) characteristics (C T Farley et al., 1998; Claire T. Farley & Morgenroth, 1999; Günther & Blickhan, 2002; Sholukha et al., 1999) and musculoskeletal system geometry (Greene & McMahon, 1979; McMahon et al., 1987).

At the whole body level, COM gravitational potential energy (E_{pot}) and mechanical kinetic energy (E_{kin}) dynamic patterns are characterized as in-phase in running (Veerle Segers et al., 2007). Specifically, both E_{pot} and E_{kin} reach minimum values at mid-stance. Further, there is minimal mechanical energy exchange between E_{pot} and E_{kin} in running (C T Farley & Ferris, 1998), due to similar fluctuation patterns of E_{pot} and E_{kin} during stance phase (Veerle Segers et al., 2007). To better understand COM mechanics patterns in running gait, it is necessary to investigate sagittal plane COM mechanical work (W_{com}) and instantaneous power (P_{com}) characteristics. Previous studies have investigated whole body COM mechanical work and power in walking and step-to-step transition (Adamczyk & Kuo, 2009; Donelan et al., 2002; Zelik & Kuo, 2010), as well as during the walk-to-run transition process (Veerle Segers et al., 2007). A greater understanding is needed regarding W_{com} and P_{com} characteristics while running across a range of speeds, as well as the effect of locomotion speed changes on COM gravitational E_{pot} and mechanical E_{kin} characteristics. Moreover, as part of the subsystem in the spring-mass model, little is known about whether there is a relationship between W_{com} and K_{vert}, and between W_{com} and K_{leg} respectively across running speeds.

The primary purpose of this study was to investigate whether change of running speeds would have influence on change of lower extremity stiffness patterns, and whole body W_{com} and P_{com} characteristics. Another goal was to investigate whether K_{vert}, K_{leg} can be predicted from K_{joint} within each running speed. Moreover, we also planned to investigate whether a connection exists between sagittal plane W^{+}_{coms} and K_{vert}, K_{leg} respectively across running speeds. Outcomes of this study would be beneficial for future lower extremity assistive device development, especially for adjustment of K_{joint} which may be used to predict K_{vert} and K_{leg}values in different running speeds. It would also be beneficial for further K_{joint} adjustment in response to different velocity change, which may result in K_{vert} and K_{leg} change. This may also be helpful to enhance running performance, since increasing passive stiffness in the musculoskeletal system influences lower extremity stiffness, which has been reported to be related to performance enhancement (Brughelli & Cronin, 2008; Lindstedt et al., 2002; Reich et al., 2000). Based on these concepts, we hypothesized that: (1) K_{vert}, W_{com} would increase with running speeds while K_{leg} would remain unchanged; (2) K_{vert} and K_{leg} would be predicted more from K_{knee}, compared with K_{ankle} and K_{hip} at each speed; (3) W_{com} would be predicted from K_{vert} and K_{leg} across running speeds.

Methods

Recruitment

Twenty abled-bodied subjects (37 $\pm$ 15.3 years, 172 $\pm$ 11.2 cm, 68 $\pm$ 14.1 kg) participated in the study. All subjects signed informed written consent approved by the university's institutional review board before participation. All subjects were without lower extremity musculoskeletal related injuries for the past 6 months before the test.

Study Design and Experimental Protocol

We measured subjects' body mass, height and leg length (L_0) before the formal test. Leg length (L_0) was measured as the vertical distance from the greater trochanter to the floor during static standing (McGowan et al., 2012). Then fifty-five retro-reflective markers were placed on the skin surface of the subjects, based on a previously published whole body marker set (Sawers & Hahn, 2012). Subjects were asked to run on a force-instrumented treadmill (Bertec, Inc., Columbus, OH) at six different speeds, from 1.8 to 3.8 m/s (0.4 m/s intervals), for 75 seconds per stage. Data were extracted from the middle strides (20 strides on average) of each stage.

Data Collection

Segmental kinematic data were collected at 120 Hz using an 8-camera motion capture system (Motion Analysis Corp., Santa Rosa, CA). Ground reaction force data were collected at 1200 Hz using the force-instrumented treadmill. Kinematic and kinetic data were filtered with a low-pass fourth-order Butterworth filter at 6 Hz and 50 Hz, respectively.

Data Analysis

Whole body COM position was calculated from weighted sum of 15-segments (head, trunk, pelvis, upper arms, lower arms, hands, thighs, shanks, and feet) full body model (Resseguie, Jin, & Hahn, 2016) for each subject in Visual 3D (C-Motion, Inc., Germantown, MD). The spring-mass model vertical stiffness (K_{vert}) was calculated from peak vertical ground reaction force ($vGRF_{peak}$) divided by vertical displacement of the COM from ground contact until mid-stance (Δy) (Fig. 6.1) (Brughelli & Cronin, 2008; C T Farley et al., 1993; Claire T Farley & Gonzalez, 1996; Ferris et al., 1998; McGowan et al., 2012; McMahon & Cheng, 1990), expressed as: $K_{vert} = vGRF_{peak}/\Delta y$. Half swept angle ($\theta$) was defined as the angle between leg-spring at ground contact and mid-stance (Fig. 6.1), and it was calculated from running speed (μ), ground contact time (t_c) and initial leg length (L_0) (Claire T Farley & Gonzalez, 1996; Ferris et al., 1998), expressed as: $\theta = sin^{-1}(\mu t_c/2L_0)$. Leg-spring maximum displacement (ΔL) can be calculated via the expression of changes in vertical COM displacement ((Δy), half swept angle (θ) and initial leg length (L_0) (Fig. 6.1) (C T Farley et al., 1993; Claire T Farley & Gonzalez, 1996; McGowan et al., 2012; McMahon & Cheng, 1990), expressed as: $\Delta L = L_0(1 - cos\ \theta) + \Delta y$. And leg stiffness ($K_{leg}$) was calculated as peak vertical ground reaction force ($vGRF_{peak}$) divided by leg-spring maximum displacement (ΔL) (Fig. 6.1) (Brughelli & Cronin, 2008; C T Farley et al., 1993; Claire T Farley & Gonzalez, 1996; Ferris et al., 1998; McGowan et al., 2012; McMahon & Cheng, 1990), expressed as: $K_{leg} = vGRF_{peak}/\Delta L$. Lower extremity joint moments were calculated using an inverse dynamics model coded in Visual 3D. Joint stiffness (K_{joint}) was calculated as a change in sagittal plane joint moment (ΔM_{joint}) divided by sagittal plane joint angular displacement ($\Delta\theta_{joint}$) in the braking phase of ground contact, based on the anterior-posterior ground reaction force value (Hobara et al., 2013; Kuitunen et al., 2002), expressed as: $K_{joint} = \Delta M_{joint}/\Delta\theta_{joint}$.

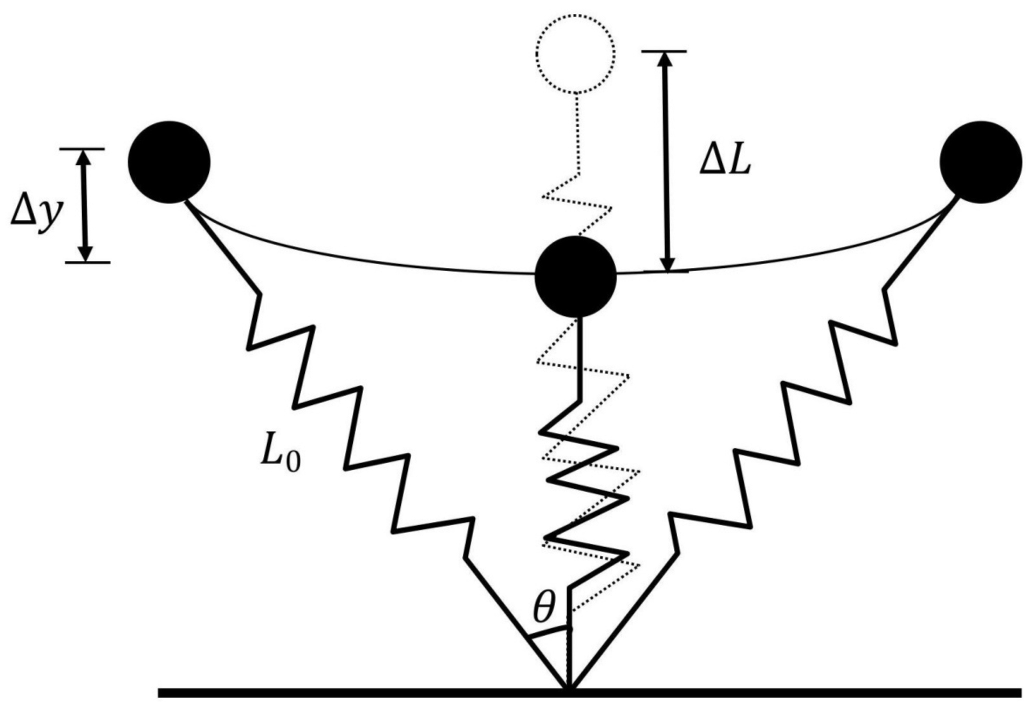

Fig 6.1. Schematic representative of a spring-mass model in running stance phase. The model consists of a point mass (COM) equivalent to body mass and the leg as a massless linear spring. Leg spring is compressed and reaches maximum compression (ΔL) at mid-stance. COM displacement in the vertical direction is denoted as Δy. Half swept angle by the leg spring is denoted as θ.

COM potential energy (E_{pot}) was calculated as the product of of body mass (M_b), gravitational constant (g = 9.81 m/s^2), and instantaneous COM height (h_i) (Veerle Segers et al., 2007), expressed as: E_{pot} = $M_b g h_i$. And COM kinetic energy (E_{kin}) was calculated from sum of E_{kin} in both horizontal and vertical direction (Veerle Segers et al., 2007), expressed as: E_{kin} = $M_b v_h^2/2 + M_b v_v^2/2$, ($v_h$ and v_v are COM velocity in horizontal and vertical direction, respectively). We also calculated COM instantaneous power in the horizontal (P_{comh}), vertical (P_{comv}) direction and sagittal plane (P_{coms}), based on the definition of a previous study (Veerle Segers et al., 2007), expressed as: P_{comh} = $M_b a_h v_h$; P_{comv} = $M_b(g + a_v)v_v$; P_{coms} = $P_{comh} + P_{comv}$; (a_h and a_v are COM acceleration in horizontal and vertical direction, respectively). Moreover, COM positive (W_{com}^{+}) and negative mechanical work (W_{com}^{-}) in the horizontal, vertical direction and sagittal plane were calculated as instantaneous positive (P_{com}^{+}) or negative power (P_{com}^{-}) in each direction or plane integrated over time, respectively (Veerle Segers et al., 2007). All outcome variables were calculated and averaged from both limbs and averaged across three gait cycles.

Ground reaction force (GRF) and virtual leg length (instantaneous leg length/L_0) force-length curve were plotted by averaging across twenty subjects (Fig. 6.2). Group mean COM potential energy (E_{pot}), kinetic energy (E_{kin}) and sagittal plane COM instantaneous power (P_{coms}) were plotted from three representative speeds (1.8, 2.6, 3.8 m/s) as well (Fig. 6.3, Fig. 6.4).

Statistical Analysis

Vertical stiffness (K_{vert}), leg stiffness (K_{leg}), joint stiffness (K_{joint}), COM positive work (W_{com}^{+}) and negative work (W_{com}^{-}) were examined using a 1-way ANOVA to compared among six speeds. Initial alpha level was set to 0.05. When main effect was detected, Bonferroni adjustments were used for pairwise comparison, so that the alpha level was divided by the number of comparisons (adjusted $\alpha = 0.0033$ for all variables' pairwise comparison in this study). Additionally, multiple linear regression analysis was conducted to develop models for predicting K_{vert}, K_{leg} from K_{joint} (ankle, knee and hip joint stiffness) within each running speed. Moreover, simple linear regression analysis was used to examine relationships between sagittal plane COM positive work (W_{coms}^{+}) and K_{vert}, K_{leg} across speeds, to investigate whether W_{coms}^{+} could be predicted from K_{vert} or K_{leg}. All statistical analyses were performed using SPSS (V22.0, IBM, Armonk, NY).

Results

Stiffness

The comparison of K_{leg} among all running speeds was not significant (p = .413). The speed main effect for K_{vert} was significant (p < .0001), so pairwise comparison was conducted: K_{vert} at 1.8 m/s was significantly lower than all speeds between 2.6 – 3.8 m/s (p < .0001); K_{vert} at 2.2 m/s was lower than all speeds between 3.0 – 3.8 m/s ($p \leq$ .0001); K_{vert} at 2.6 m/s was lower than at 3.4 m/s (p = .001) and 3.8 m/s (p = .0002); and K_{vert} at 3.0 m/s was lower than at 3.8 m/s (p = .0032) (Table 6.1). For K_{joint} comparison, speed main effect was significant in K_{knee} (p < .0001), and pairwise comparison was conducted: K_{knee} at 1.8 m/s was lower than at 3.0 m/s (p = .002) and 3.8 m/s (p = .001); K_{knee} at 2.2 m/s was lower than at 3.0 m/s (p = .001) and 3.8 m/s (p = .003) (Table 6.1)

Table 6.1. Vertical stiffness (KN/m), leg stiffness (KN/m) and joint stiffness (Nm/kg/deg) across running speeds. Sample Mean (SD); n = 20.

	Running Speed (m/s)					
	1.8	**2.2**	**2.6**	**3.0**	**3.4**	**3.8**
K_{vert}	23.03 (5.19)[a]	24.98 (4.77)[b]	27.10 (4.50)[a,c]	29.79 (4.70)[a,b,d]	32.84 (6.40)[a,b,c]	40.29 (9.16)[a,b,c,d]
K_{leg}	13.49 (3.40)	13.39 (3.85)	13.22 (3.28)	13.07 (2.76)	12.96 (3.65)	13.45 (4.17)
K_{ankle}	0.18 (0.08)	0.18 (0.05)	0.19 (0.06)	0.19 (0.09)	0.21 (0.07)	0.23 (0.09)
K_{knee}	0.10 (0.02)[e]	0.11 (0.02)[f]	0.12 (0.03)	0.14 (0.04)[e,f]	0.15 (0.06)	0.18 (0.08)[e,f]
K_{hip}	0.25 (0.14)	0.22 (0.11)	0.26 (0.12)	0.24 (0.07)	0.27 (0.10)	0.27 (0.10)

a: Statistically significant differences of K_{vert} between 1.8 m/s and all speeds between 2.6 – 3.8 m/s, respectively ($p <$ .0001);
b: Differences of K_{vert} between 2.2 m/s and all speeds between 3.0 – 3.8 m/s, respectively ($p \leq$ .0001);
c: Differences of K_{vert} between 2.6 m/s and 3.4 m/s (p = .001), 2.6 m/s and 3.8 m/s (p = .0002);
d: Differences of K_{vert} between 3.0 m/s and 3.8 m/s (p = .0032);
e: Differences of K_{knee} between 1.8 m/s and 3.0 m/s (p = .002), 1.8 m/s and 3.8 m/s (p = .001);
f: Differences of K_{knee} between 2.2 m/s and 3.0 m/s (p = .001), 2.2 m/s and 3.8 m/s (p = .003).

Mechanical Work

Speed main effects were significant in both W_{coms}^{+} (p < .0001) and W_{coms}^{-} (p = .002), so pairwise comparison was conducted: W_{coms}^{+} at 1.8 m/s was lower than at 3.0 m/s (p = .002), 3.4 m/s (p < .0001) and 3.8 m/s (p = .003) (Table 6.2); and W_{coms}^{-} at 1.8 m/s was lower than at 3.4 m/s (p = .002) (Table 6.2). Speed main effects were also significant in both W_{comh}^{+} (p < .0001) and W_{comh}^{-} (p < .0001), and pairwise comparison was conducted: W_{comh}^{+} at 1.8 m/s was lower than all speeds between 2.6 – 3.8 m/s (p < .0003), W_{comh}^{+} at 2.2 m/s was lower than all speeds between 3.0 – 3.8 m/s (p < .002), W_{comh}^{+} at 2.6 was lower than at 3.4 m/s and 3.8 m/s, respectively (p < .001), W_{comh}^{+} at 3.0 m/s was lower than at 3.8 m/s (p = .0009); and W_{comh}^{-} at 1.8 was lower than at all speeds between 2.6 – 3.8 m/s (p < .0001), W_{comh}^{-} at 2.2 m/s was lower than at 3.4 m/s (p = 0.0004) (Table 6.2).

Table 6.2. Whole body COM positive and negative mechanical work (J/kg) in sagittal plane, horizontal and vertical direction across speeds. Sample Mean (SD); n = 20.

	Running Speeds (m/s)					
	1.8	**2.2**	**2.6**	**3.0**	**3.4**	**3.8**
W^{+}_{coms}	1.03 (0.14)[a]	1.06 (0.23)	1.16 (0.14)	1.21 (0.20)[a]	1.22 (0.31)[a]	1.31 (0.29)[a]
W^{-}_{coms}	0.85 (0.11)[b]	0.90 (0.12)	0.96 (0.11)	0.96 (0.13)	0.98 (0.15)[b]	0.94 (0.19)
W^{+}_{comh}	0.21 (0.05)[c]	0.26 (0.08)[d]	0.33 (0.09)[c,e]	0.39 (0.12)[c,d,f]	0.43 (0.17)[c,d,e]	0.54 (0.17)[c,d,e,f]
W^{-}_{comh}	0.17 (0.05)[g]	0.22 (0.05)[h]	0.30 (0.08)[g]	0.33 (0.08)[g]	0.37 (0.09)[g,h]	0.39 (0.12)[g]
W^{+}_{comv}	0.83 (0.14)	0.81 (0.17)	0.85 (0.12)	0.84 (0.13)	0.81 (0.16)	0.79 (0.18)
W^{-}_{comv}	0.69 (0.11)	0.69 (0.11)	0.68 (0.10)	0.65 (0.11)	0.62 (0.10)	0.56 (0.11)

Note: COM negative mechanical work data were presented in absolute values.
a: Statistically significant differences of W^{+}_{coms} between 1.8 m/s and 3.0 m/s (p = .002), 1.8 m/s and 3.4 m/s (p < .0001), 1.8 m/s and 3.8 m/s (p = .003);
b: Differences of W^{-}_{coms} between 1.8 m/s and 3.4 m/s (p = .002);
c: Differences of W^{+}_{comh} between 1.8 m/s and all speeds between 2.6 – 3.8 m/s, respectively (p < .0003);
d: Differences of W^{+}_{comh} between 2.2 m/s and all speeds between 3.0 – 3.8 m/s, respectively (p < .002);
e: Differences of W^{+}_{comh} between 2.6 m/s and all speeds between 3.4 – 3.8 m/s, respectively (p < .001);
f: Differences of W^{+}_{comh} between 3.0 m/s and 3.8 m/s (p = .0009);
g: Differences of W^{-}_{comh} between 1.8 m/s and all speeds between 2.6 – 3.8 m/s, respectively (p < .0001);
h: Differences of W^{-}_{comh} between 2.2 m/s and 3.4 m/s (p = .0004).

Multiple and Simple Linear Regression

Results from the multiple linear regression analysis showed that K_{joint} could predict K_{vert} at 1.8 m/s and 2.2 m/s (Table 6.3). At 1.8 m/s, the model accounted for 38.4% of the variance in K_{vert} (R^2 = 0.384, p = .046), and K_{knee} made the strongest unique contribution to predict K_{vert} at this speed (β = 0.509, p = .022) (Table 6.3). At 2.2 m/s, the model accounted for 49.8% of the variance in K_{vert} (R^2 = 0.498, p = .014), and K_{knee} again made the strongest unique contribution to predict K_{vert} at this speed (β = 0.553, p = .011) (Table 6.3).

Additionally, K_{joint} could predict K_{leg} among most speeds, except at 3.0 m/s and 3.8 m/s (Table 6.3). At 1.8 m/s, the model accounted for 42.4% of the variance in K_{leg} (R^2 = 0.424, p = .028), and K_{knee} made the strongest unique contribution to predict K_{leg} (β = 0.532, p = .014) (Table 6.3). At 2.2 m/s, the model accounted for 79.3% of the variance in K_{leg} (R^2 = 0.793, p < .0001). For this speed however, K_{knee} (β = 0.553, p = .0004) and K_{hip} (β = 0.526, p = .001) both made strong unique contributions to predict K_{leg} (Table 6.3). At 2.6 m/s, the model accounted for 39.9% of the variance in K_{leg} (R^2 = 0.399, p = .039), and K_{knee} made a unique contribution to predict K_{leg} (β = 0.456, p = .04) (Table 6.3). At 3.4 m/s, the model accounted for 47.4% of the variance in K_{leg} (R^2 = 0.474, p = .026), and K_{hip} made a strong unique contribution to predict K_{leg} (β = 0.721, p = .009) (Table 6.3).

Simple linear regression analysis showed that K_{leg} could not predict W^{+}_{coms} across speeds (R^2 = 0.133, p = .477). However, K_{vert} could significantly predict W^{+}_{coms} across speeds (R^2 = 0.902, p = .004) with strong positive association between K_{vert} and W_{coms}^{+} (r = 0.95) (Table 6.3).

Table 6.3. Multiple linear regression models between joint stiffness and vertical stiffness, leg stiffness respectively at statistical significant speeds; Simple linear regression model between vertical stiffness and whole body COM sagittal plane positive work across speeds (n = 20), marked in grey shading.

Variable	**Model**					
	$\boldsymbol{\beta_0}$	$\boldsymbol{B}$	$\boldsymbol{\beta}$	$\boldsymbol{R^2}$	$\boldsymbol{p}$ **- value**	$\boldsymbol{\sigma_{est}}$
K_{vert} *at 1.8 m/s*						
Model Summary	8.298			0.384	0.046	4.438
K_{ankle}		15.078	0.246		0.301	
K_{knee}*		106.818	0.509		**0.022**	
K_{hip}		5.236	0.142		0.549	
K_{vert} *at 2.2 m/s*						
Model Summary	9.289			0.498	0.014	3.705
K_{ankle}		3.927	0.040		0.832	
K_{knee}*		109.190	0.553		**0.011**	
K_{hip}		14.566	0.338		0.093	
K_{leg} *at 1.8 m/s*						
Model Summary	4.815			0.424	0.028	2.809
K_{ankle}		-3.062	-0.076		0.736	
K_{knee}*		73.127	0.532		**0.014**	
K_{hip}		7.793	0.323		0.169	
K_{leg} *at 2.2 m/s*						
Model Summary	3.210			0.793	< 0.0001	1.921
K_{ankle}		-18.779	-0.237		0.065	
K_{knee}*		88.051	0.553		**< 0.001**	
K_{hip}*		18.308	0.526		**0.001**	
K_{leg} *at 2.6 m/s*						
Model Summary	4.512			0.399	0.039	2.772
K_{ankle}		2.576	0.048		0.826	

K_{knee}*		45.443	0.456		**0.040**	
K_{hip}		10.603	0.404		0.071	
K_{leg} *at 3.4 m/s*						
Model Summary	9.760			0.474	0.026	2.920
K_{ankle}		-18.732	-0.353		0.128	
K_{knee}		3.013	0.046		0.835	
K_{hip}*		25.250	0.721		**0.009**	
K_{vert} *predict* W^{+}_{coms}						
Model Summary	0.677	0.016	0.950	0.902	0.004	0.038

*Statistically significant contribution of joint stiffness to predict the models are indicated in bold. $\boldsymbol{\beta_0}$**:** linear regression model constant (*y* intercept); $\boldsymbol{B}$**:** unstandardized coefficients; $\boldsymbol{\beta}$**:** standardized coefficients; $\boldsymbol{\sigma_{est}}$**:** standard error of the estimate.

Curve Patterns

Based on the stance phase ground reaction force and virtual leg length relationship for three representative speeds, we found that the slope of the curve increased as running speeds increased, and virtual leg length magnitude tended to decrease (Fig. 6.2). The COM gravitational E_{pot} remained relatively unchanged among three representative running speeds while the magnitude of E_{kin} increased dramatically when speeds increased (Fig. 6.3). Lastly, the magnitude of sagittal plane COM instantaneous power (P_{coms}) tended to increase as running speeds increased (Fig. 6.4).

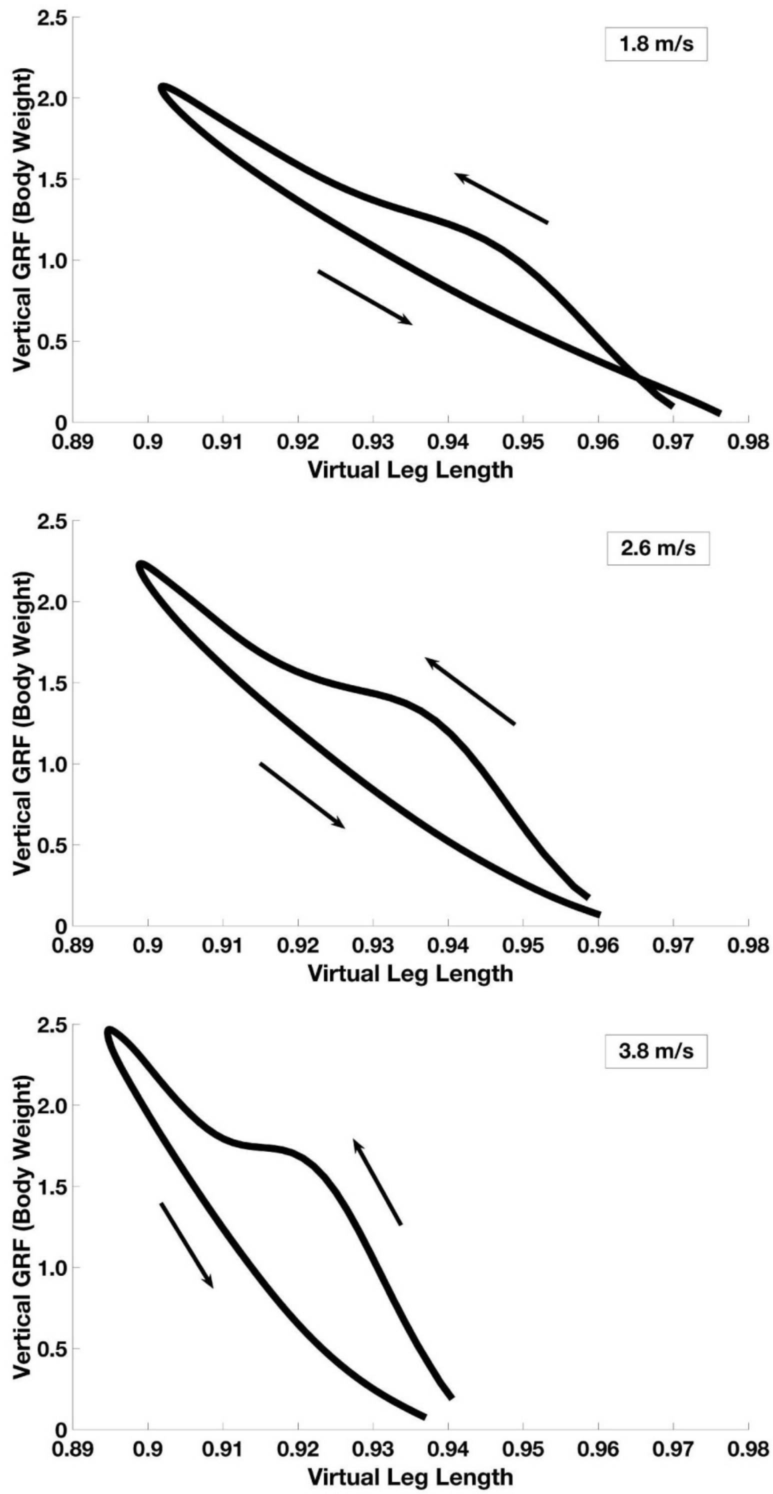

Fig 6.2. Group average (n = 20) leg-spring force-length curve at three representative speeds. GRF: vertical ground reaction force normalized to body weight; Virtual leg length: instantaneous leg length/L_0.

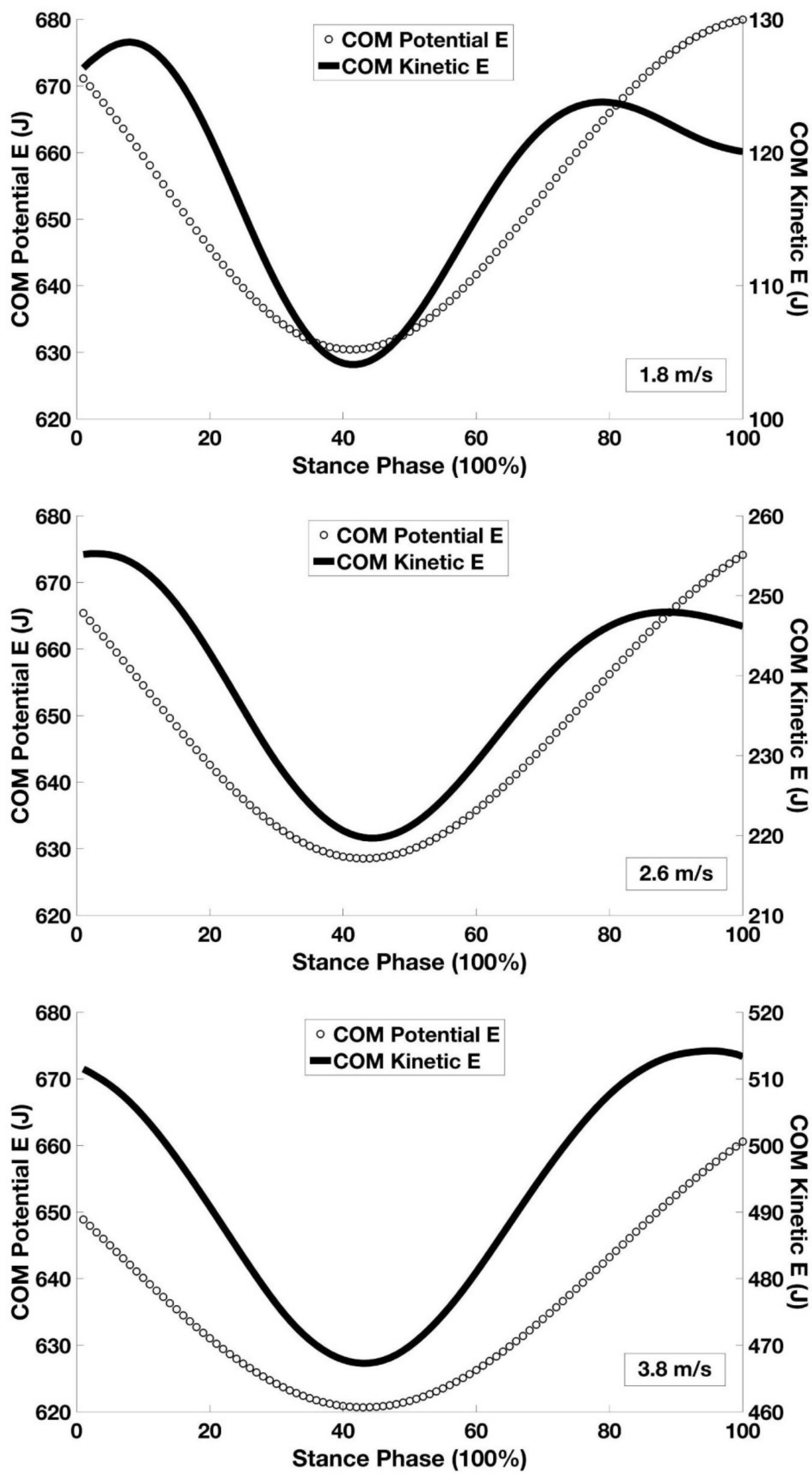

Fig 6.3. Group average (n = 20) whole body COM gravitational potential energy (E_{pot}) and mechanical kinetic energy (E_{kin}) in stance phase of three representative speeds.

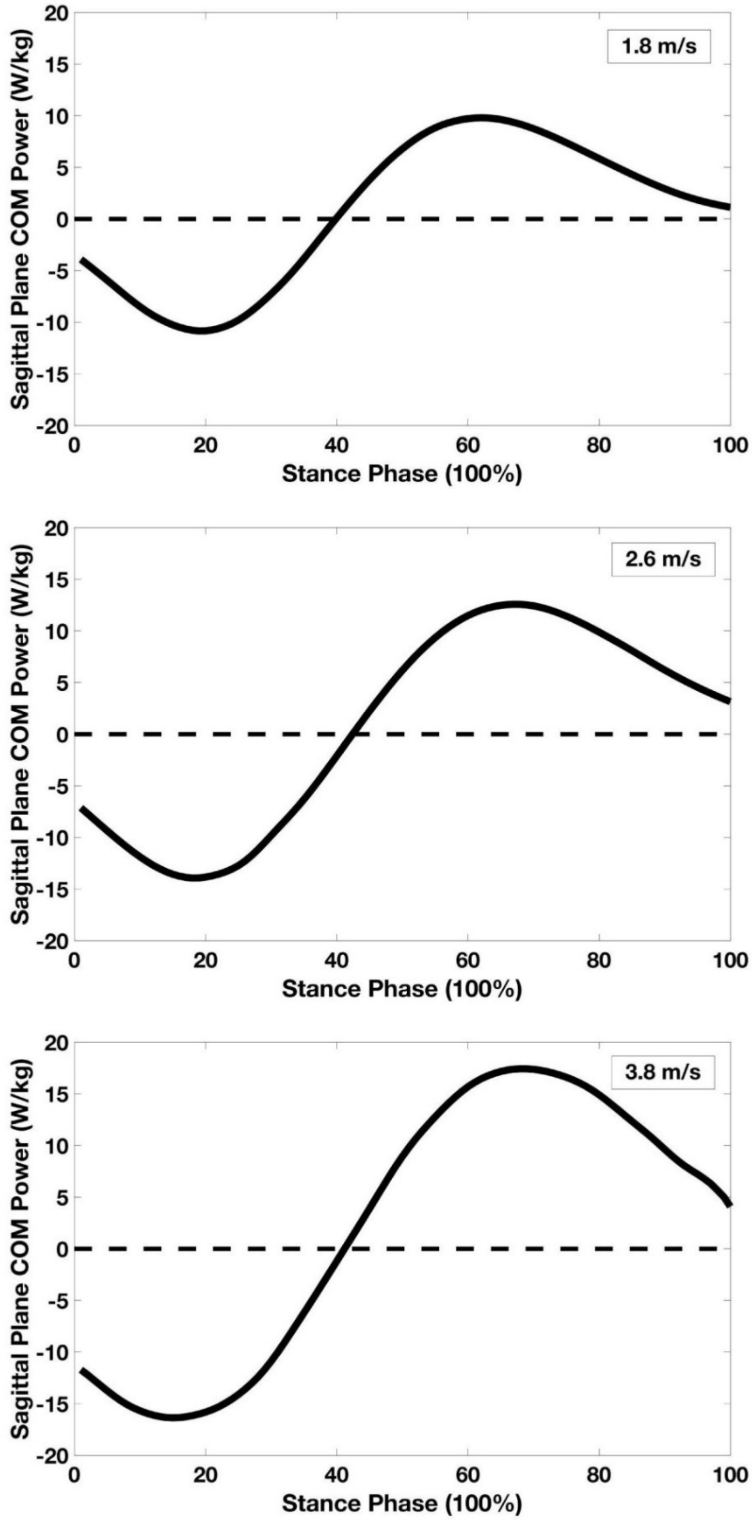

Fig 6.4. Group average (n = 20) sagittal plane whole body COM instantaneous mechanical power (P_{coms}) at stance phase of three representative speeds.

Discussion

The primary goal of this study was to investigate whether vertical stiffness (K_{vert}) and leg stiffness (K_{leg}) can be predicted from lower extremity joint stiffness (K_{joint}). Specifically, we aimed to investigate whether K_{ankle}, K_{knee} and K_{hip} could predict K_{vert} and K_{leg} using multiple linear regression models for each running speed. Additionally, we also investigated whether sagittal plane COM positive work (W^{+}_{coms}) was associated with K_{vert} or K_{leg}; specifically, whether W^{+}_{coms} could be predicted from K_{vert} or K_{leg} across running speeds. The initial hypothesis that K_{knee} would have a larger contribution to predict K_{vert} and K_{leg} was supported. The hypothesis that W^{+}_{coms} could be predicted from K_{vert} and $K_{leg}$was partially supported.

Both K_{vert} and K_{leg} could be predicted from K_{joint} in the multiple linear regression models at slow speeds (1.8 and 2.2 m/s) (Table 6.3). Further, K_{knee} made a significant unique contribution to predict K_{vert} and K_{leg} at these speeds (Table 6.3). However, K_{vert} could not be predicted from K_{joint} among speeds from 2.6 – 3.8 m/s. One reason may be that K_{vert} tended to increase as running speeds increased, Interestingly, the change of running speeds had mixed effects on K_{joint}: K_{knee} tended to increase while K_{ankle} and K_{hip} fluctuated more as running speed increased (Table 6.1). The other reason may be attributed to the observation that K_{vert} is more related to whole body COM bouncing oscillation patterns (Brughelli & Cronin, 2008; G. Cavagna et al., 1988; McMahon et al., 1987), and it seems that K_{joint} would have a closer relationship with leg-spring stiffness rather than with COM oscillation characteristics. For multiple linear regression analysis between K_{leg} and K_{joint}, K_{knee} and K_{hip} made larger contributions to predict K_{leg} (Table 6.3). Both K_{knee} and K_{hip} contributed to predict K_{leg} at 2.2 m/s (Table 6.3). K_{hip} made a unique contribution to predict K_{leg} at 3.4 m/s (Table 6.3). Interestingly, it seems that K_{ankle} did not make much contribution to predict K_{leg} across all running speeds in this study. However, K_{knee} and K_{hip} both made large contributions, especially K_{knee} made the largest contribution to predict K_{leg} among most speeds (Table 6.3). This may be attributed to the idea that the human leg is a system comprised of multiple springs and the sub-springs can be coordinated with each other during ground contact in running. Under similar loading conditions, the spring with smallest stiffness will undergo the largest displacement and this would have the most influence on the overall leg-spring system stiffness (Claire T. Farley & Morgenroth, 1999). In this study, K_{knee} tended to be lower than K_{ankle} and K_{hip} across all running speeds (Table 6.1). Besides making the largest contribution to predict K_{leg} among speeds, knee joint flexion (relatively lower stiffness) could also be beneficial for elastic energy storage in the first half of running stance phase and the following energy return in the second half of stance (Jin & Hahn, 2018; Kuitunen et al., 2002). In the simple linear regression analysis, K_{vert} and W_{coms}^{+} had a strong positive association across running speeds. This may be due to both K_{vert} and W_{coms}^{+} tending to increase as running speeds increased; the other reason would be attributed to both variables being closely related to COM dynamics and mechanical energy characteristics in running stance phase.

The other goal of the study was to examine whether change of running speeds would have effects on K_{vert}, K_{leg} and W_{com}. The initial hypothesis was partially supported. Results showed that K_{vert} increased with running speeds while K_{leg} remained relatively unchanged. These findings were in agreement with previous findings (Biewener, 1989; Brughelli & Cronin, 2008; G. A. Cavagna, 2005; C T Farley et al., 1993; He et al., 1991; McMahon & Cheng, 1990; Morin et al., 2005, 2006). Change of speeds had effects on both positive and negative W_{com} in sagittal plane and horizontal direction (Table 6.2). However, change of speeds did not have significant effects on either positive and negative W_{com} in vertical direction. This finding can be explained by sagittal plane COM instantaneous power curve characteristics, as well as COM gravitational E_{pot} and mechanical E_{kin} curve patterns (Fig. 6.3, Fig. 6.4). The magnitude of peak sagittal plane P^{+}_{coms} and P^{-}_{coms} tended to increase as running speeds increased (Fig. 6.4), and the area below the curve (W^{+}_{coms} and W^{-}_{coms}) increased as well. Among the three representative running speeds, both maximum and minimum E_{pot} values did not change much, while the magnitude dramatically increased for E_{kin} as running speeds increased (Fig. 6.3). This indicates that change of running speeds has more effects on E_{kin} than on E_{pot}. Further, E_{kin} was more sensitive to speeds change in the horizontal direction than in the vertical direction as running speeds increased. Additionally, GRF increased in both vertical and horizontal direction as speeds increased, indicating that the COM energy absorption was greater in the first half of stance and higher speeds also required more positive work generated on the COM to assist the body to move forward (horizontal direction) in the following propulsion period. This helps explain why W^{+}_{comh} and W^{-}_{comh} increased as speeds increased. Moreover, the magnitude for E_{pot} tended to decrease when speeds increased (Fig. 6.3). This may be attributed to the observation that COM displacement in the vertical direction decreases as running speeds increase (Brughelli & Cronin, 2008). Reducing COM maximal height would be beneficial for maintaining whole body COM dynamic system stability, as well as optimization of energy transfer between E_{pot} and E_{kin} when locomotion task demand is increased.

We also investigated vertical GRF and virtual leg length relationship in three representative speeds (Fig. 6.2). The curve consisted of an ascending and a descending phase. The ascending phase represents the loading period and the descending phase represents the unloading period. Within the ascending phase, the "yielding" pattern became more obvious as speeds increased (Fig. 6.2). Additionally, virtual leg length at initial contact decreased as speeds increased, indicating that the leg-spring compressed more as speeds increased. This would be beneficial for energy absorption as external impact force increases, and it could also be beneficial for reducing COM height and E_{pot} as speeds increases. Moreover, the magnitude of virtual leg length change tended to decrease as speeds increased (Fig. 6.2). This indicate that the leg-spring becomes stiffer as running speeds increased.

Limitations

One limitation of this study is that the leg spring was assumed to not be compressed at initial ground contact in the spring-mass model. As speeds increased, the initial leg length was lower than static standing leg length (L_0), which was used in the K_{leg} calculation. This likely affected K_{leg} results at relatively higher speeds. Additionally, a treadmill running protocol was used in this study, with controlled locomotion speeds and thus some individual variations may have been constrained. Another limitation is that we only investigated slow to medium range of running speeds.

Whether COM dynamic patterns would be different in a wider range of speeds requires further investigation.

Future Work

Future studies should compare the accuracy of different models in predicting COM dynamic patterns during locomotion. In this study, we calculated COM instantaneous mechanical power from kinematic variables of COM movement (COM velocity and acceleration). The method was proved to be reliable in estimating COM displacement compared with the method derived from GRF (Veerle Segers et al., 2007). Other studies have used dot product of GRF and COM velocity to estimate COM external mechanical power, and COM velocity was derived from integration of GRF in these studies (G. A. Cavagna, 1975; Donelan et al., 2002; Zelik & Kuo, 2010). Further comparison between these two methods in both walking and running across speeds is needed. Moreover, it would be interesting conduct a simulation analysis to investigate the optimization of whole body COM dynamic characteristics, by adjusting lower extremity kinematic and kinetic variables in both walking and running, to enhance gait performance.

Conclusion

When running from slow to medium speeds, leg spring stiffness remains relatively unchanged while vertical stiffness tended to increase with speeds; whole body COM positive and negative mechanical work tended to increase in both sagittal plane and in horizontal direction. Both leg stiffness and vertical stiffness could be predicted from lower extremity joint stiffness in the multiple linear regression models at 1.8 m/s and 2.2 m/s. Leg stiffness could be predicted at a wider range of running speeds from joint stiffness, compared with vertical stiffness. The knee joint contributed more to predict vertical stiffness and leg stiffness. Sagittal plane COM positive work could be predicted from vertical stiffness and the two variables had a strong positive association when running speeds increased. These findings suggest that leg spring system stiffness could be predicted from subsystem joint level stiffness characteristics. Lastly, whole body COM mechanical work had a strong positive association with COM oscillation patterns in stance phase running across different speeds.

CHAPTER VII

CONCLUSION

Summary of Results and Findings

This dissertation investigated lower extremity joint level kinematic and kinetic characteristics in both walking and running gait across speeds, as well as in gait transition processes. First, change of locomotion tasks and speeds effects on lower extremity joint kinetic patterns were investigated in Chapter III. Next, progression of age effects on lower extremity joint level kinematic and kinetic characteristics, and general gait patterns were examined in Chapter IV. Then lower extremity joint kinetic patterns during the gait transition processes between walking and running were investigated in Chapter V. Lastly, in Chapter VI the investigation was expanded beyond the lower extremity system to investigate the potential connections with whole body COM dynamic and mechanical patterns.

Findings from Chapter III indicate change of locomotion speeds significantly affect joint level kinetic characteristics within both walking and running locomotion states. The ankle joint was determined to be critical during stance phase for energy generation in both walking and running across different speeds. Higher ankle joint stiffness was associated with more positive work performed and power generation in running. Additionally, different locomotion task demands (walking vs. running) could fundamentally change lower extremity joint level kinetic patterns, even at the same locomotion speed.

Chapter IV revealed some differences of joint level kinematic and kinetic characteristics between young and middle age groups in walking and running across different speeds. Specifically, the middle age group had higher ankle plantar flexor moment angular impulse, higher total lower extremity support torque and higher hip joint positive work in the stance phase of walking. The middle-age group also exhibited higher knee flexion angle at initial ground contact in walking and running. These findings indicate that progression of age can affect ankle and hip joint kinetic patterns across walking speeds, knee joint kinematic patterns in both walking and running, and some spatiotemporal changes in running speeds.

Chapter V demonstrated that switching gait patterns between walking and running would have significant effects on lower extremity joint kinetic patterns. In stance phase, an energy generation and transfer phenomenon occurred between distal and proximal joints during both WRT and RWT processes. The energy generation and transfer direction was opposite between WRT and RWT. The stance phase mechanical energy generation transfer mechanism was sensitive at the transition step (S0) for both WRT and RWT. Moreover, joint power patterns and ankle joint angle-moment curve characteristics at the transition step were similar to target locomotion patterns in WRT. These results extended the knowledge framework about the effects of locomotion speed and task changes on lower extremity joint mechanics patterns.

Finally, the results from Chapter VI indicate a connection exists between whole body COM oscillation patterns and lower extremity joint level kinetic characteristics in running, and COM mechanical work had a positive association with COM oscillation patterns in stance phase across speeds. Additionally, compared with vertical stiffness, leg stiffness can be predicted at a wider range of running speeds from joint stiffness. These findings have built a connection between the whole body COM system and the lower extremity system and expand the perspective for future study of gait efficiency optimization and performance enhancement.

The findings of these studies help to further clarify the effects of locomotion task and speed changes on lower extremity joint kinematic and kinetic characteristics, and the connection between whole body COM dynamic patterns with lower extremity joint mechanical characteristics. Very broadly, this work can serve as a reference for future foot-ankle system assistive device development, and potential optimization of whole body COM dynamic patterns across speeds. As such, future work should investigate lower extremity kinematic and kinetic characteristics between healthy subjects and patient populations over a wider range of age and locomotion speeds. Additionally, further investigation of whole body COM dynamic patterns and the connection with the lower extremity system in gait transition processes is necessary. Further, simulations to explore the optimization of COM mechanics characteristics and gait performance enhancement would be another future goal, based on the current findings from these studies. Overall, this work demonstrates the potential connection between COM dynamic patterns and lower extremity system kinetic characteristics, and lays a knowledge framework for future gait efficiency optimization and performance enhancement by adjusting lower extremity joint level kinematic and kinetic variables. However, some limitations may have affected the results of this study and therefore more work is needed prior to potential assistive device development and performance enhancement.

Limitations

One limitation of the whole study is inherent to treadmill walking and running. As the locomotion speeds and acceleration magnitudes were controlled, different subjects may have individual variations which may have been masked by the controlled speeds, and the results may be different from the naturally occurring patterns of over-ground locomotion and gait transition. Additionally, we only selected one representative acceleration and deceleration speed value for WRT and RWT in Chapter V. Different acceleration and deceleration magnitudes would likely affect gait transition speed and transition step gait patterns (L. Li, 2000). This suggests that the results from the current transition study may not represent gait transition patterns with other acceleration and deceleration speeds.

Second, we assumed gait symmetry between left and right legs in all locomotion conditions. We recruited young and middle-age healthy subjects in the study, and all subjects were without any musculoskeletal injuries which may affect locomotion at least six months before the test. However, some subjects may have relied more on one leg in walking and running. In cases where there is pronounced asymmetry, the individual distribution of joint work and power may vary beyond what was observed in the present study.

Third, the spring-mass model used in Chapter VI assumed that the leg spring was not compressed at initial ground contact in running. Our observations showed that the initial leg length was measured to be less than static standing leg length (L_0) as running speeds increased. This may have affected K_{leg} results at higher running speeds.

Recommendations for Future Work

It is apparent that some additional work is needed before applying the knowledge to assistive device development, which could be suitable for multiple locomotion tasks across speeds for patient populations, as well as potential optimization of gait efficiency and performance enhancement.

Firstly, future work should compare lower extremity kinematic and kinetic patterns between patient populations (lower limb amputees, spinal-cord injury patients, etc.) with current non-injured subjects. This would help to further identify differences of gait patterns and joint mechanics characteristics between different population groups and settings, especially the compensatory mechanisms within the lower extremity musculoskeletal system among patient populations in both walking and running across speeds. Additionally, for current assistive device sensors and intelligent control algorithm development, further investigation and exploration is needed. One goal would be to develop sensors which can more accurately detect locomotion speeds or task changes, and then adjust foot-ankle system stiffness via control algorithms, resulting in control parameters that are more suitable for a wider array of speeds and locomotion task demands. Such work would be beneficial for development of foot-ankle system assistive devices suitable for walking and running across a range of speeds, and better assisting with gait transition for patient populations.

Another avenue of future work based on the current findings would be optimization of whole body COM dynamic patterns across different locomotion speeds. Based on the findings of Chapter VI, COM movement mechanics exhibited connection with lower extremity system kinetic patterns. Via simulation analysis approaches, a more thorough understanding could be established about how to adjust lower extremity kinematic and kinetic variables to optimize COM dynamic patterns, which could be beneficial for optimization of gait efficiency and performance enhancement.

With these potential areas of work being carried out and the respective challenges overcome, the future is bright for assistive device development and gait efficiency optimization.

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