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KINETIC ASYMMETRIES DURING SUBMAXIMAL
AND MAXIMAL SPEED RUNNING

A Thesis Presented

by

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ABSTRACT

KINETIC ASYMMETRIES DURING SUBMAXIMAL AND MAXIMAL SPEED RUNNING

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The ability to attain maximal running velocity is a major factor in many athletic events. Typically human sprinters increase center of mass (CoM) velocity at submaximal speed by increasing both stride length and stride frequency, but then rely predominantly on increasing stride frequency to attain maximal CoM velocity (Luhtanen and Komi, 1978). An important issue for sports scientists, coaches and athletes is an understanding of the factors within a stride that can enhance or limit maximal running speed. Previous research has identified horizontal braking and propulsive impulses, vertical propulsive impulse, effective vertical stiffness, peak concentric hip extension muscle power during swing and peak eccentric knee flexion muscle power during swing as potential kinetic limiters of maximal CoM velocity (Cavagna, 2006; Chapman and Caldwell, 1983b; Korhonen et al., 2008; Weyand et al., 2001). Bilateral asymmetry is an observed difference in kinetic or kinematic gait parameters between the right and left sides (Zifchock et al., 2006), and is present for many of these parameters at maximal speed (Belli et al., 1995; Exell et al., 2012). Bilateral differences in any of these parameters could negatively affect an athlete's ability to increase their stride frequency, but the degree to which such asymmetries change as running speed progresses from submaximal

to maximal is unknown. It was hypothesized that asymmetries in key sprinting parameters would be larger at maximal speed than all other tested speeds. Nine female competitive speed and power athletes (age = 21 ± 3 years, mass = 60.58 ± 7.48 kg, height = 1.64 ± 0.07 m) completed two trials at each of 70%, 80%, 90% and 100% of their maximal speed on a force-instrumented treadmill. A repeated-measures ANOVA was completed for each parameter to examine the asymmetry differences across speed. The only parameter for which asymmetry was statistically greater ($p < 0.05$) during maximal speed than all other speeds was effective vertical stiffness, in which the level of asymmetry increased incrementally with speed ($r^2 = 0.97$). Therefore the hypothesis that asymmetries in general would increase with speed for all key parameters is rejected. Bilateral asymmetries in effective vertical stiffness appeared to be related to asymmetries in both vertical and A/P propulsive impulse at maximal speed. Furthermore, asymmetries in effective vertical stiffness may force runners to resort to a less stable and less coordinated gait, limiting their ability to further increase stride frequency, and thus limiting maximal speed.

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CHAPTER 1

INTRODUCTION

1.1 Background

The ability to attain maximal running velocity is a major factor in many athletic events. During a single gait cycle, rotational motions of the limbs and trunk are converted into translational motion of the centre of mass (CoM). Typically human sprinters increase CoM velocity at submaximal speed by increasing both stride length and stride frequency, but then rely predominantly on increasing stride frequency to attain maximal CoM velocity (Luhtanen and Komi, 1978). Thus to sprint faster, athletes must progress through both stance and swing phases more quickly. An important issue for sports scientists, coaches and athletes is an understanding of the factors within a stride that can enhance or limit maximal running speed.

Neglecting wind and air resistance, the ground reaction force (GRF) during stance is the only external force responsible for changing an athlete's forward speed. Therefore runners can increase CoM velocity by applying greater forces to the ground. Not surprisingly, faster athletes produce higher propulsive GRF in the sagittal plane (Korhonen et al., 2008; Brughelli, Cronin and Chaouaci, 2010). However, with increases in stride frequency, the athlete has less time to do so due to a shorter stance phase. It has been shown that effective vertical impulse (the product of foot-ground contact time and the vertical force exceeding the body's weight) decreases to a minimum as runners attain maximal speed (Weyand, 2010). Further, vertical impulse during the braking phase of early stance has been negatively correlated to sprint speed (Belli, 1995; Cavagna, 2006). Therefore, a speed performance limitation could be the compromised ability of an

athlete's leg extensor muscles to produce enough force to increase the propulsive anterior and vertical GRF and impulse under the time constraint of a reduced contact phase.

As increased forces over decreased contact times are required of the lower limb musculature to attain high sprinting velocities, vertical stiffness characteristics of the contact limb could limit maximal sprint speed (Cavagna, 2006; Weyand et al., 2010). There is an optimal stiffness for both the minimization of contact time and maximization of power output of the leg muscles during sprinting activities (Flanagan, 2007); however leg effective vertical stiffness increases as a function of stride frequency (Farley, 1996). Consequently, vertical stiffness increases associated with high stride frequencies could be sub-optimal, and thus have detrimental and limiting effects on sprint performance. Leg stiffness is regulated through muscular force production, and a recent modelling study indicated that the force-velocity relationship limits maximal sprint speed more than other muscle mechanical properties (Miller, Umberger and Caldwell, 2012). Although the exact details of this limitation within the gait cycle were not elucidated, muscular limitations to sprint speed could occur during both the stance and swing phases.

The posterior leg muscles are active in late stance for forward propulsion and to initiate a rapid swing phase. They are also important in late recovery to help position and orient the foot correctly for proper force application without increasing the posteriorly directed braking impulse. At high speeds the recovery process must be completed in a shorter amount of time, requiring rapid changes of lower limb segmental energy. Through muscular efforts the leg segments move relative to each other, changing the energy level of the lower limb segments (Robertson and Winter, 1980). The total lower limb energy needs to increase to initiate recovery, but then must be decreased to correctly reposition

the limb for the next stance phase; both processes can be completed faster with more powerful muscle contractions (Chapman & Caldwell, 1983a). Lower limb energy can be increased by powerful hip extension in early swing. Later in recovery the hamstring muscle group eccentrically produces force to decrease limb energy for foot placement. High limb energy means faster limb recovery, so the later this eccentric energy reduction the better. However, too much limb energy in late swing can promote excessive posteriorly-directed braking impulse that will slow down the runner during the beginning of the subsequent stance phase. Therefore, the hamstrings must act powerfully to quickly decrease limb energy late in swing (Chapman and Caldwell, 1983a). The ability of an athletes' lower limb muscles to rapidly raise and lower total limb energy to recover and correctly initiate the next stance phase have been implicated as potential limiters of sprint speed (Chapman & Caldwell, 1983b).

Locomotion is a cyclical action modulated by interactions of the body's neural and mechanical dynamic systems (Masani, Kouzaki and Fukunaga, 2000). These system interactions lead to natural variability in the movements that produce forward progression. Many authors believe that excess variability implies system instability that could be a factor in preventing effective locomotor performance. Furthermore, several authors have postulated that the interactions between the stance and swing phases may contribute to limiting maximal velocity (Weyand et al, 2000; Weyand et al, 2010; Chapman and Caldwell, 1983b; Miller, Umberger and Caldwell, 2012). As a consequence, it is believed that gait variability should be kept as low as possible (Masani, Kouzaki and Fukunaga, 2000). Inherent intra-limb variability between strides has been shown to increase with speed (Belli et al., 1995; Zifchock, 2006), indicating that running

gait may be compromised in moving a runner's CoM forward at higher speeds, limiting maximal performance. In contrast, dynamical systems theory argues that task variability is inherently functional because we do not have a singular predefined motor program by which a task can be completed. In fact healthier and more robust systems are characterized by containing an increased number of unique methods of completion for a single task (Bernstein, 1967). In this view, increased variability could lead to better performance at higher speeds. While it is clear that the effects of intra-limb differences on performance are not well understood, what about the effects of between-limb differences?

Bilateral asymmetry is an observed difference in kinetic or kinematic gait parameters between the right and left sides (Zifchock et al., 2006), and has been the subject of many investigations during walking (Sadeghi et al., 1997) and submaximal running (Zifchock et al., 2008). Researchers have mostly evaluated the degree of bilateral asymmetry by comparing unilaterally-collected data from non-consecutive steps (Giakas, 1997). As with intra-limb variability, it has been demonstrated that as stride frequency increases sprinters' limbs may behave in an increasingly asymmetrical manner (Belli et al., 1995). However, under the same conditions increased inter-limb variability has been reported for various temporal and kinematic parameters (Belli et al., 1995; Karamanidis, Arampatzis and Bruggemann, 2003). Therefore, as maximal speed is approached each stride becomes less similar to both the previous and subsequent strides, complicating asymmetry analyses. Unfortunately, differences between limbs may not be significant if the intra-limb variability is too high (Exell, 2012). Clearly, the view of asymmetry is clouded by the intra- and inter-limb variability inherent in athletes.

The relationship between stride frequency and inter-limb differences has not been well-established in runners, although increased variation between right versus left hip and knee angular velocities at stride frequencies greater than 10% of preferred has been reported (Karamanidis, Arampatzis and Bruggemann, 2003). Belli et al. (1995) reported increased asymmetry in both contact time and center of mass vertical displacement as speed increased from submaximal to supra-maximal. However, the bilateral differences were not statistically significant due to high inter-limb variability in these measures at top speeds. Swanson and Caldwell (2005) noticed significant bilateral differences in peak concentric hip extension, and eccentric knee flexion powers in 5 of 8 subjects as they approached maximal speed. In addition, a study of eight national level sprinters revealed that at their highest stride frequency, every athlete exhibited unique kinetic asymmetries, although they did not necessarily have high kinematic asymmetries (Exell et al., 2012). Although both kinematic and kinetic gait asymmetries are reported in the literature, it is important to focus on the biomechanical parameters associated with the causes rather than the effects of motion. Maximal speed will be limited by underlying kinetic capabilities rather than the kinematic motions that they produce. Although kinematic asymmetries may aid interpretation, the real problem will be centered on kinetic asymmetries during high speed velocity running.

1.2 Statement of Problem and Hypothesis

How does speed influence asymmetry? Although kinematic and kinetic bilateral asymmetries have been reported during maximal speed running, the literature is sparse regarding the asymmetry characteristics of runners as they approach their maximal speed. Attainment of maximal velocity requires increasing stride frequency, and the present

study proposes to examine the consequences of continually decreasing stance and swing time on inter-limb differences. Therefore, key kinetic parameters identified as potentially limiting to maximal velocity will be evaluated for asymmetry at maximal and three sub-maximal (70%, 80% and 90% of maximal) sprinting speeds on a force treadmill. We hypothesize that these kinetic parameters will exhibit greater bilateral asymmetry at maximal speed than at all tested submaximal speeds.

1.3 Significance of the Study

This study will add to the previous work on asymmetry and maximal velocity running. Although asymmetry has been investigated during walking, submaximal and maximal speed running (Allard et al., 1996; Zifchock et al., 2008; Exell et al., 2012), the changes in asymmetry of kinetic variables implicated in limiting maximal speed have not been examined during progressions from submaximal to maximal speed sprinting. Exploring these asymmetry changes with increasing speed is necessary to understand the potential contribution of asymmetry to sprint performance. It is likely that the mechanical limitations of sprint performance are athlete-specific, and there is evidence indicating bilateral asymmetries at maximal velocity follow this trend (Exell, 2010). Determining the result of increasing speed on inter-limb differences may yield information about asymmetry's role during sprint performance. Asymmetry has also been viewed as a clinical indicator of injury (Exell, 2010; Zifchock et al., 2006). Therefore examining asymmetry during high velocity running may give clinicians a different perspective as to why injuries may occur, and what effects those injuries may have on an athlete's performance.

CHAPTER 2

REVIEW OF LITERATURE

The ability to attain maximal running velocity is a major factor in many athletic events. In track events specifically, regardless of the distance, the single most important performance characteristic is speed (Seagrave, 1996). Maximal velocity attainment has been the focus of much previous research, which will form the initial portion of this literature review. After a discussion of step characteristics and their interactions with respect to speed, this review will examine limitations to maximal sprint speed during both the stance and swing phases. Finally, attention will be focused on the effects of bilateral asymmetry on maximal speed.

2.1 The Gait Cycle

Sprint running is a repeated series of gait cycles, where the rotational motions of the limbs and trunk are converted into translational motion of the total body centre of mass (CoM). A stride is one complete gait cycle, defined by the time elapsed from an initial foot contact to the subsequent foot contact of the same lower limb (Figure 1).

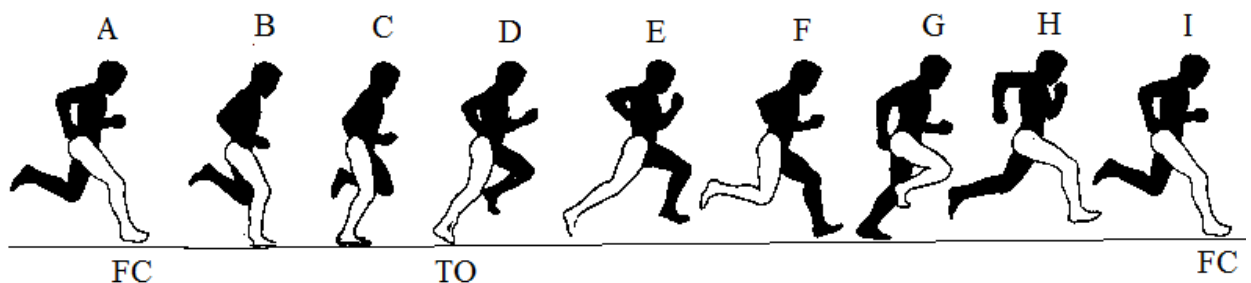


Figure 2.1. Depiction of one full stride. Initial foot contact (FC) is the instant the white limb comes into contact with the ground, while toe off (TO) is the instant it leaves the ground.

A single stride encompasses stance (Figure 2.1 A-D) and swing phases (Figure 2.1 D-I) of both legs, although the right and left leg phases are shifted in time with respect to each

other. During stance phase, the athlete's foot is on the ground and can apply forces to the surface, unlike swing phase when the athlete's foot is off the ground. The duration for which a limb is in contact with the ground is called contact time (Figure 2.1 A-D). Flight time refers to the period during which neither limb is in contact with the ground (Figure 2.1 D-F). A step is the time elapsing one foot contact to the foot contact of the ipsilateral limb, approximately one half of a stride (Figure 2.1 A-F). Stride length (SL) refers to the distance travelled during a cycle, while stride frequency (SF) quantifies how many occur per second. Stride length is measured in units of meters (m), while stride frequency is measured in cycles per second (Hz or s^{-1}). The product of SL and SF (or step length and step frequency) yields stride velocity (SV) measured in units of meters per second (ms^{-1}). Definitions of strides and steps are inconsistent in the literature. Not surprisingly, the absolute values of cycle lengths and frequencies could be drastically different between studies due to the discrepancies in definitions. The important point is how these parameters interact to attain top speed, and the pattern is the same whether we discuss strides or steps.

2.2 Attaining top Speed

Maximizing the translational velocity of the CoM relies on maximizing SV. As SV is the product of SF and SL, an intuitive guess would be that peak SV will occur via the simultaneous maximization of both SF and SL. Paradoxically though, as an athlete approaches maximal velocity, an increase in one parameter is typically countered by a decrease in the other (Bosco, 1986). SF wholly depends on step and stride time, while SL depends on stance and flight distance and these characteristics are inherently different (Hunter, 2004). Using members of the 1987 Italian National Sprints Team as subjects a

pictorial representation of the optimal combination of SF and SL was developed (Figure 2.2), highlighting that neither maximizing SF nor SL actually yields top speed (Donati, 1995).

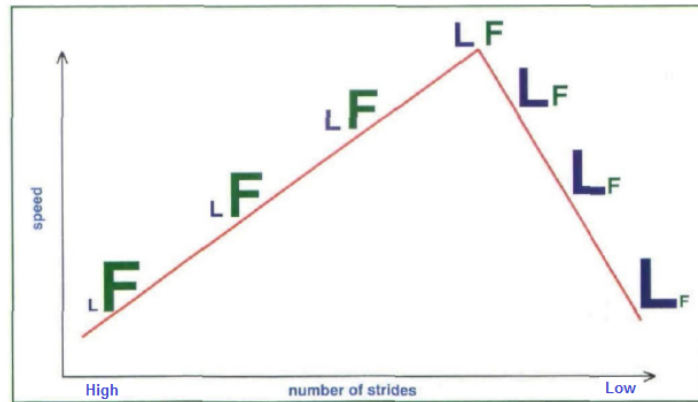


Figure 2.2. Relationship between Stride Frequency and Stride Length. Stride Frequency (SF) and Stride Length (SL) with respect to speed. Adapted from Donati, 1995.

The large and small ‘F’ and ‘L’ denote SFs and SLs above and below optimal respectively. Top speed (SV) is reached with the correct combination of SF and SL, and any deviation from optimal resulted in a decrease in speed. Therefore, it is the optimization rather than the maximization of both SF and SL that allows SV to reach its peak. To attain top speed the widely accepted theory is that sprinters will first increase SL to increase speed at submaximal levels, and then increase SF to approach their highest speeds (Figure 2.3) (Luhtanen and Komi, 1978; Weyand et al., 2000; Mero and Komi, 1986; Kuitunen et al., 2002).

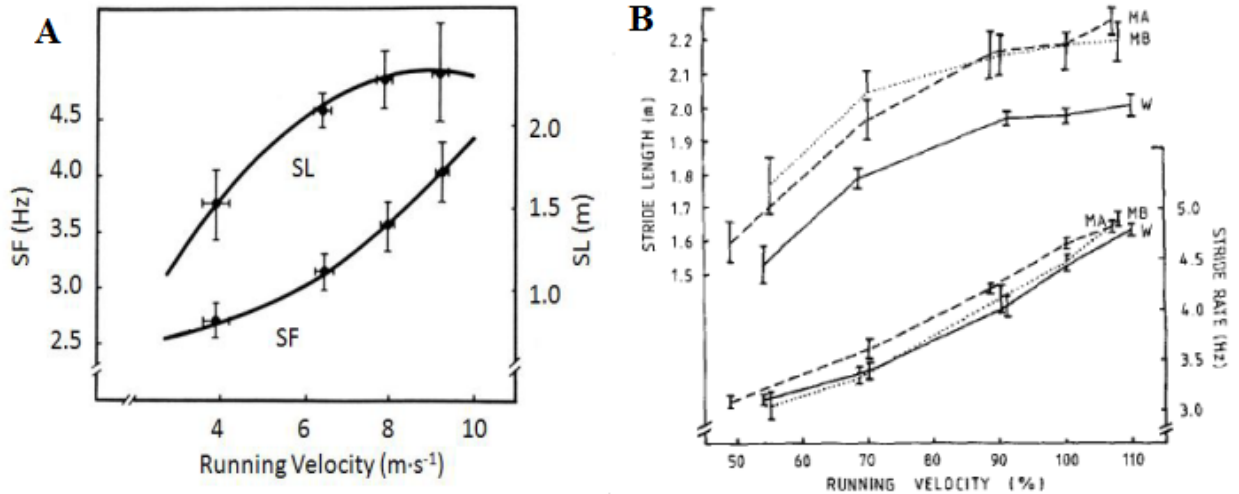


Figure 2.3. Athlete-Specific Stride Length and Frequency. A) Step Frequency (SR) and Step Length (SL) at various measured running speeds. SF values are on the left vertical axis and SL values are on the right vertical axis. Adapted from Luhtanen and Komi in Asmussen and Jorgensen, 1978. B) Mean (\pm SD) stride rates (SF) and SLs of top male sprinters (MA), less-skilled male sprinters (MB) and female sprinters (W) at running velocities ranging from 50% to 110% maximal. Adapted from Mero and Komi, 1986.

However, individual athletes will have unique optimal combinations of SF and SL, mainly due to anatomical differences (Donati, 1995; Kunz and Kaufmann, 1981). Thus, the exact timing and magnitude of increases in SF and SL will also differ among athletes (Figure 3b). While individual strategies to increase speed are variable, the overall trend to attain maximal speed is to increase SF, indicating that each phase of the gait cycle needs to occur in less time.

2.3 Mechanical Energy During Stance

Running faster requires the generation of more mechanical energy (Appendix A). Briefly, an increase in mechanical energy (positive mechanical work) will increase the velocity of a system. If our mechanical system consists of the runner and the immediate running environment, the transfer of energy between the runner and environment can only take place when the runner is in contact with the ground. As a runner presses on the ground, the ground pushes back with an equal and opposite ground reaction force (GRF),

which accelerates (either positively or negatively) the CoM of the runner. The amount of work done on an object directly influences changes in mechanical energy. Related is the mechanical impulse, the product of force and the time the force was applied to the object; applied impulse cause changes in momentum and velocity.

2.4 Impulse and Stiffness in Sprint Running

Horizontally, impulses are categorized as either propulsive or braking. Braking impulse occurs in early stance when force application is directed anteriorly, resulting in a posterior GRF acting in opposition to the CoM velocity, slowing the sprinter down. Propulsive impulse occurs in later stance when a sprinter applies a posterior force to the ground, resulting in an anteriorly directed GRF, accelerating the runner's CoM in the direction of travel. Higher velocities are achieved when the braking and propulsive impulses are minimized and maximized respectively (Hunter, Marshall and McNair, 2005). To enter the flight phase a sprinter's foot must leave the ground, a result of vertical impulse. As the sprinter presses into the ground, the GRF will act to accelerate the runner's CoM in a vertical direction. Applying greater forces in opposition to gravity increases a runner's vertical velocity on takeoff, flight time and forward distance traveled between steps (Weyand et al., 2001).

It has been suggested that the lower limb can behave like a spring to allow humans and animals to run at faster speeds (Heglund et al., 1982). In order to minimize muscular work during the stance phase, spring-like behavior of lower limb tissues could store energy during the initial portion of ground contact, to be released later in stance to contribute to vertical and horizontal GRFs (Farley and Gonzalez, 1996). Although there are many tissues (muscles, bones, ligaments, tendons) that can behave as springs, a single

linear spring-mass model of the lower limb as presented in Figure 4 has been used to represent all of the lower limb structures during stance phase (Farley and Gonzalez, 1996).

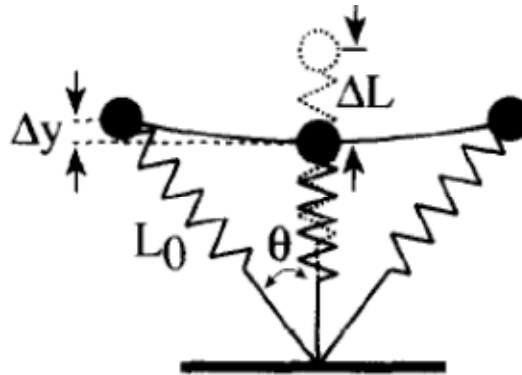


Figure 2.4. The Spring-Mass Model during Running. The black dot is the mass representative of the center of mass of the runner. L_0 is the initial length of the leg spring, also represented in a standing situation by the dotted outline of the spring-mass model. ΔL is the difference between L_0 and the length of the leg spring at maximal compression caused by the peak compressive force (F). Δy represents the downward vertical displacement of the center of mass during the stance phase. θ represents half the angle swept by the spring-mass model during ground contact. For an overview of the measurements and calculations of the different parameters of the model, refer to Appendix B. Adapted from Farley and Gonzalez, 1996.

This spring-mass model works on the premise that the leg is an inverted compressible spring attached to the CoM; the spring-mass unit rotates as an inverted pendulum as the runner progresses through stance. This model represents running conceptually because the spring allows the CoM height to be at a minimum when the hip of the stance leg passes over the ankle (mid-step) (McMahon and Cheng, 1990). If the model were to have no spring, the CoM would be highest at mid-step; representing walking (McMahon and Cheng, 1990). Mid-step tends to be the point at which athletes produce peak vertical GRF, as they switch from applying braking to propulsive impulse (McMahon and Cheng, 1990).

Mechanical stiffness, the relationship between the peak compressive force and the deformation experienced by a spring, can be used to quantify the ability of a spring to absorb and release mechanical energy. Stiffness of a spring is described in terms of its spring constant (k) derived from force (F) and displacement (x) in Equation 1.

$$\begin{aligned} F &= k \cdot x \\ k &= F/x \end{aligned} \quad (1)$$

Thus force (F) applied to an end of a spring is the product of stiffness k and displacement x from its equilibrium position. A high k value means that the spring is stiff, having the ability to absorb, store and release mechanical energy quickly in response to an applied force (Bret et al., 2002). This simple mechanical characteristic of springs could be advantageous during the maximal velocity phase of sprinting when the contact times are the shortest. Not surprisingly many authors have studied mechanical stiffness of the lower limb during sprinting (e.g. Bret et al., 2002; McMahon and Cheng, 1990; Chelly and Denis, 2000; Farley et al., 1993). Many categories of mechanical stiffness have been described in the literature such as *leg stiffness* (McMahon and Cheng, 1990, Farley and Gonzalez, 1996); *joint stiffness* (Arampatzis et al., 1999); *musculotendinous, tendon and passive stiffness* (reviewed in Brughelli and Cronin, 2008); and *effective vertical stiffness* (EK_{vert}) (McMahon and Cheng, 1990; Morin et al., 2005; Bachman, Heise and Bressel, 1999; Farley and Gonzalez, 1996). Leg stiffness refers to the ratio of force to the displacement of the CoM for a given point in time (Farley, Glasheen and McMahon, 1996), shown in Figure 4 as $(F/\Delta L)$. Joint stiffness is calculated by the ratio of joint moment to angular joint displacement (Arampatzis, 1999), while musculotendinous, tendon and passive stiffness have been calculated with oscillation techniques, ultrasonography and isokinetic dynamometers respectively (Brughelli, 2008). EK_{vert}

represents the ratio of peak vertical force to the peak vertical displacement of the center of mass; $F/\Delta y$ in Figure 4. EK_{vert} describes the vertical motions of the center of mass during the entire ground contact time, and is determined by a combination of the stiffness of the leg spring, the angle swept by the leg spring and the compression of the leg spring ($F/\Delta L$, θL and ΔL respectively from Figure 4). McMahon and Cheng (1990) found that EK_{vert} increases quadratically with speed across many species of bipedal and quadrupedal hoppers and runners (Figure 5).

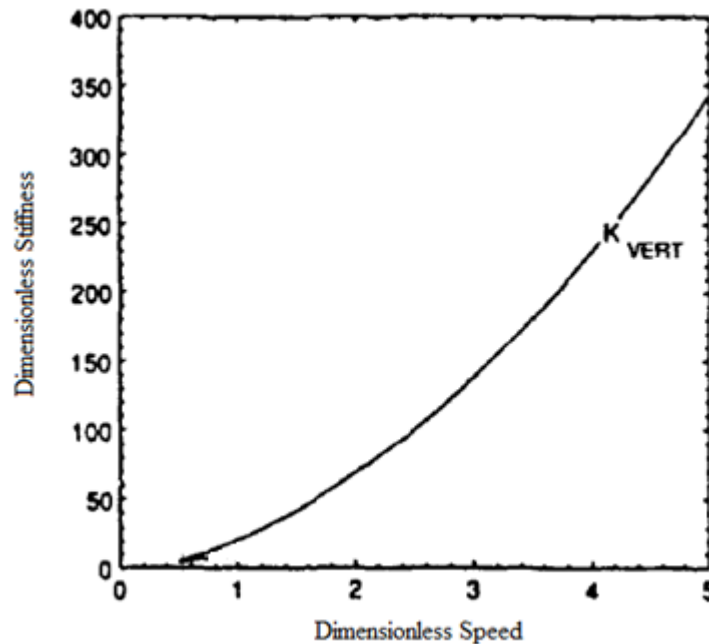


Figure 2.5. Effective Vertical Stiffness and Speed Relationship. EK_{vert} (K_{vert}) versus speed adapted from McMahon and Cheng, 1990. Both speed and stiffness values are dimensionless because the authors were comparing animals of different sizes (including humans).

Some authors have used kinematic data to determine the change in leg length during the stance phase of running (Arampatzis, Bruggemann and Metzler, 1999; Luhtanen and Komi, 1980; Mero and Komi, 1986). Arampatzis and colleagues compared computer-modelled (equations from McMahon and Cheng, 1990) and kinematically-derived values of EK_{vert} and leg stiffness of 13 runners at five velocities ranging from

2.5m/s to 6.5m/s. The authors found that at all speeds the modelled vertical leg stiffness values were significantly less than the kinematically-derived values. However, the EK_{vert} values were the same regardless of the method of computation (Arampatzis, Bruggemann and Metzler, 1999). The authors concluded that the difference in leg stiffness values were due to an overestimation of the change in leg length by the model. “Elasticity of the support leg” has also been used to describe the combined elasticity of the mechanical system of man during the support phase, and is evaluated using a spring constant as well (Luhtanen and Komi, 1980; Mero and Komi, 1986). These authors also used kinematic data to determine the change in leg length during the stance phase. The authors split the stance phase into two components; the time before the lowest point of the center of mass during contact (eccentric phase) and the time after lowest point (concentric phase). EK_{vert} can increase with either decreased CoM displacement or increased force applied to the ground. Both of these occur as speed increases (McMahon and Cheng, 1990). This is beneficial because stiffer springs spend less time compressing, limiting the amount of time the runner will spend on the ground during stance phase.

Clearly, while the foot is on the ground, changes in the mechanical energy of the whole body are of importance. However, when the body is airborne, energy changes within the limb become priority (Elftman, 1938). When gravity is the only force acting on an object, the object’s overall mechanical energy level cannot be changed. Therefore, it is imperative that the runner’s leg be re-connected with the ground as fast as possible, but also in the correct position for proper force application. This can be accomplished by continually altering the mechanical energy profile of the swinging lower limb.

2.5 Mechanical Energy During Swing

Changing the mechanical energy within the limb is a result of mechanical work performed by muscles (Robertson and Winter, 1980; Caldwell and Forrester, 1992). Appendix C outlines a method based on the early work of Robertson and Winter (1980) for calculating instantaneous segmental energy, work and power (Caldwell and Forrester, 1992). Net positive segmental power indicates that energy is being generated to the segment, while net negative power signifies energy removal. (Robertson and Winter, 1980). These are determined by the summation of all joint force and muscle moment powers acting on a segment (Appendix C). In this type of analysis, a major assumption is that ligaments, bone forces, and friction play no part in energy generation or absorption, so only the sign of the joint force and muscle moment powers dictate whether energy is being produced or absorbed by a segment. In other words, any movement of the joint is only attributed to muscular force production, thus segments will “generate” or “absorb” energy during swing phase because of muscles (Robertson and Winter, 1980). Eccentric and concentric force production by muscles absorbs energy from, and generates energy to the segment respectively (Robertson and Winter, 1980). The phenomenon of energy transfer is also illustrated in Appendix C. When both segments happen to be rotating in the same direction there will be energy transfer. The direction of the transfer of energy will depend on the relative angular velocities and joint moments of the segments. Energy will transfer from the segment with the low angular velocity to the segment with the high angular velocity regardless of concentric or eccentric muscular force production. When the net difference in angular velocities between segments is 0 (as in isometric force production), energy will be transferred from the segment with a negative moment power

(joint moment in the opposite direction to the angular velocity) to the segment with a positive muscle moment power (joint moment in the same direction as angular velocity). The idea of energy transfer within a limb is quite abstract, and it would be advantageous to add anatomical significance to the simple models of energy transfer.

Aleshinsky (1986) described eleven possible “power sources” for each segment, responsible for energy generation, absorption, and transfer within the segment. Six of the eleven sources are analogous to PM and PJ as described by Robertson and Winter, 1980 (occurring at each segment end), with PJ broken up into its vertical and horizontal components. The seventh source is due to gravity, while the remaining sources relate to the difference between the velocities of the segment’ end points and the centre of mass in the vertical and horizontal directions (Aleshinsky, 1986). Aleshinsky suggested that calculating segmental energy using different combinations of his “power source” approach would be more accurate than the “classic” methods of energy calculation (such as those proposed by Robertson and Winter, 1980). Caldwell and Forrester, 1992 attempted to validate Aleshinsky’s theory, but found that different power source approaches may still yield the same segmental energy value for a given segment. The authors questioned the need of a more complicated model, when the results were comparable to simpler ones. Caldwell and Forrester (1992) related the Aleshinsky equations to four types of energy transfer: joint, tendon, whip, and pendulum. Joint and tendon energy transfers occur between segments through the joint center and muscle/tendon complex, respectively. Both simultaneously increase the energy level of one segment while decreasing the energy level of the adjacent segment. Joint transfer was referred to as joint power, while tendon transfer was described as one possible part of

muscle power by Robertson and Winter (1980). Intrasegmental (within segment) energy transfer can occur in two ways; by converting rotational energy into translational energy via whip transfer, calculated from the differences between segmental endpoint and centre of mass angular velocities, and by transferring potential energy into vertical kinetic energy (pendulum transfer). Joint transfer is responsible for the majority of the energy transfer during gait, followed by whip transfer. Pendulum and tendon transfer are responsible for a limited amount of energy transfer, and may be gait-specific (Caldwell and Forrester, 1992).

2.6 The Pattern of Mechanical Energy During Swing

Muscles generate, absorb and transfer energy to and from different segments, altering the energy profile in the lower limb during swing phase. Figure 6 depicts the changes in lower limb segmental and total energy (TLE) during swing phase in running (Chapman and Caldwell, 1983a).

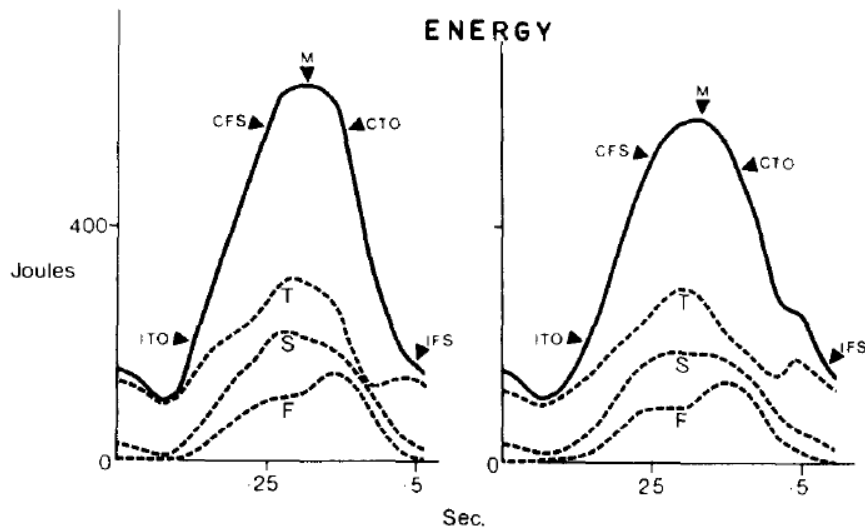


Figure 2.6. Total limb energy during swing. TLE (thick black line) and thigh (T), shank (S), and foot (F) energy changes during swing phase. Swing phase is between ITO (ipsilateral toe off) and IFS (ipsilateral foot strike).

As in stance, high levels of energy relate to high forces and velocities, so an increase in TLE occurs during the first half of swing to increase the forward velocity of the limb. This is followed by a decrease in TLE before heel strike to slow the relative forward limb motion; properly positioning the foot for ground placement and reducing the braking impulse (Hunter, Marshall and McNair, 2005). The leg musculature must complete mechanical work to change TLE, indicating that swing is not simply a passive phase (Marsh, 2004; Heglund, 1982).

2.7 Stance Phase Limitations of a Faster Gait Cycle

With higher speeds and stride frequencies, overall stride time decreases. During stance, the runner can increase CoM velocity through the application of ground reaction force (GRF) and impulse in the sagittal plane. To increase speed, the athlete must increase the propulsive anterior GRF and impulse but with reduced contact time. Pressing harder (and faster) into the ground requires more mechanical work, and the inability to increase mechanical work production further eventually limits maximal speed (Cavagna, Thys and Zamboni, 1971).

Faster sprinters take less time to complete each stance phase than slower sprinters (Hunter, 2004). Indeed, age-related declines in sprint speed are partially attributable to increases in contact time (Korhonen, 2008). In addition, as an athlete is towed at a supra-maximal speed faster than they can run unaided, their contact times decrease further (Mero and Komi, 1985). Proceeding through stance phase faster affects horizontal and vertical velocities of the CoM, as well as EK_{vert} .

Korhonen and colleagues (2008) determined that slower, older runners had propulsive GRFs smaller than similar younger and faster runners. Propulsive GRF has

been positively correlated to maximal running velocity in young and old experienced sprinters ($r=0.44$; Korhonen et al., 2008), and Australian Rules Rugby players ($r=0.47$; Brughelli, Cronin and Chaouaci, 2010). The duration of the GRF is also important, as researchers have shown that the duration of the propulsive push phase limits sprint speed (Cavagna, Komarek and Mazzoleni, 1971). The duration of the push phase is significantly longer than the brake phase up to speeds of 14 km/hr, but not from 15-20 km/hr (Cavagna, 2006). In a study of 6 national and international track and field power athletes, contact time was significantly less at the 95% maximum velocity sprint condition than any other submaximal condition on a high-speed treadmill (Kivi, Maraj and Gervais 2002). When sprinters are pulled by a rope to attain supramaximal velocity, their contact times decrease but the shank is further in front of the center of mass, indicating increased braking (Mero and Komi, 1985; Bosco and Vittori, 1986). This implies that the propulsive push duration compensates for the increased braking to attain supramaximal velocity. However, since the rope has a pull of 30-45 N, the athlete may not have to propel himself as much as simply avoid falling. Clearly, both the force of propulsion (Korhonen et al., 2008; Brughelli, Cronin and Chaouaci, 2010) and the duration of the propulsive force (Cavagna, 2006; Cavagna, Komarek and Mazzoleni, 1971; Kivi, Maraj and Gervais, 2002; Mero and Komi, 1985; Bosco and Vittori, 1986) have an impact on maximal velocity attainment. Because the product of force and time is impulse, it is safe to say that propulsive horizontal impulse is a limiting factor in an athlete's maximal sprint speed.

Braking time does not decrease nearly as much as push time as speed increases from <5km/hr to >20km/hr, demonstrating that braking is a necessary component of the

sprint motion (Cavagna, 2006). Braking forces must be kept as low as possible to avoid the loss of forward velocity in early stance (Mero and Komi, 1986). In fact, researchers reported that one runner lost 35% of their kinetic energy due to braking impulse (Cavagna, Komarek and Mazzoleni, 1971), and that the forces experienced during this phase increase with speed (Mero and Komi, 1986). Braking forces were negatively correlated with braking time across runners of all ages ($r = -0.44$) indicating that as contact times decrease, braking force increases (Korhonen et al., 2008). This might limit the amount that the sprinter could increase SF due to the increased demands imposed by these braking forces. Braking time has also been negatively correlated to maximal velocity in experienced runners ages 17-82 ($r = -0.65$) (Korhonen et al., 2008), implying that faster speeds are achieved by minimizing the time spent braking. Kinematically, this is evident from the touchdown distances of the contact limb ahead of the CoM in maximal and supramaximal sprint conditions (averaging $0.24 \pm 0.04\text{m}$ and $0.26 \pm 0.04\text{m}$ respectively) for 22 male and female sprinters (Mero and Komi, 1986). Relative braking impulse explained 7% of the variance in sprint speeds in 28 male university students, with a weak trend for faster runners to produce lower braking impulses ($R^2 = 0.04$; Hunter, Marshall and McNair, 2005). These results must be viewed with caution, because acceleration, not maximal velocity was tested due to the short 25m running track.

While propulsive GRFs are thought to be maximized simply by greater hip extension velocities, the generation of braking GRFs is more complex. They are thought to be reduced by minimizing the forward horizontal velocity of the foot at touchdown, thus allowing immediate rapid hip joint extension and knee joint flexion in early stance (Hunter, Marshall and McNair, 2005). Another important feature is to minimize the

distance the foot is placed in front of the center of mass at the instant of touchdown (Hunter, Marshall and McNair, 2005). Horizontal GRFs are integral to forward motion of a sprinter, but to enter the aerial phase of gait a sprinter must leave the ground, a result of vertical GRF production.

At any speed, applying greater forces in opposition to gravity would increase a runner's vertical velocity on takeoff, thereby increasing both the aerial time and forward distance traveled between steps (Weyand et al., 2001). A study of 33 physically active male and female runners, revealed that the vertical GRF to weight ratio was 1.26 times greater for a runner with a top speed of 11.1 m/s compared to 6.2 m/s (Weyand et al., 2001). Another study by the same group found that subjects had the ability to produce higher peak GRFs than were observed at maximal speed (Weyand et al., 2010). In response to their results the authors concluded that the limit to speed is reached when foot-ground contact times and vertical impulses decrease to the minimums that provide just enough aerial time to reposition the swing limb for the next step (Weyand et al., 2010). They related these findings to the force-velocity relationship of the contractile element of muscle, in that the faster the muscle is shortening, the less force it can produce. In this way, the decreased ability to produce vertical GRFs to propel a sprinter into swing phase could potentially limit maximal sprint speed.

It has been established that runners can increase lower limb stiffness to store and release energy at higher speeds (Figure 5) and stride frequencies (Farley and Gonzalez, 1996). Not surprisingly, significantly higher peak spring constants are observed in elite as compared to less-skilled sprinters (35.29 ± 6.64 and 14.42 ± 2.40 N/mm/kg respectively) (Mero and Komi, 1986). These results indicate that higher-skilled sprinters are better at

reusing stored energy during stance than unskilled sprinters. EK_{vert} calculated during hopping tasks has also been correlated to maximal sprint speed ($r=0.68$) in handball players (Chelly and Denis, 2000). Bret et al. (2002) determined that higher EK_{vert} was more closely associated with maximal running velocity than with the acceleration or deceleration phases of a 100m sprint. EK_{vert} was also found to increase ~3.5 fold across stride frequency changes of -26% to +36% preferred SF (Farley and Gonzalez, 1996). EK_{vert} increases as speed and SF increase, and is perhaps important to effectively absorb, store and release energy to keep the center of mass moving forward with high velocity. EK_{vert} is a measure of general limb stiffness and in the opinion of this author differences seen in EK_{vert} will have more of an impact on maximal velocity than differences seen in other, more specific measures of stiffness. Previous work has indicated that there may be an optimal stiffness for maximizing power output of the lower limb muscles (Arampatzis et al., 2001), thus any potential SF-related increase in stiffness may actually compromise lower limb power output.

2.8 Swing Phase Limitations of a Faster Gait Cycle

The implication for swing phase is that the lower limb must be repositioned in an advantageous position for force application during the next stance phase, again in a shorter amount of time. Because of the mechanical interdependence between the stance and swing phases of the stride, the vertical forces and impulses required to attain any speed are largely dependent on how rapidly the limbs can be repositioned (Weyand et al., 2001). As running speed increases, both the magnitude and the changes of mechanical energy of the lower limb increase (Willems, Cavagna and Heglund, 1995). The segmental kinetic patterns produced by muscular activity give insight as to the muscle groups

responsible for the pattern of energy flow in the lower limb during swing. These kinetic patterns are described by net joint moments (Figure 7).

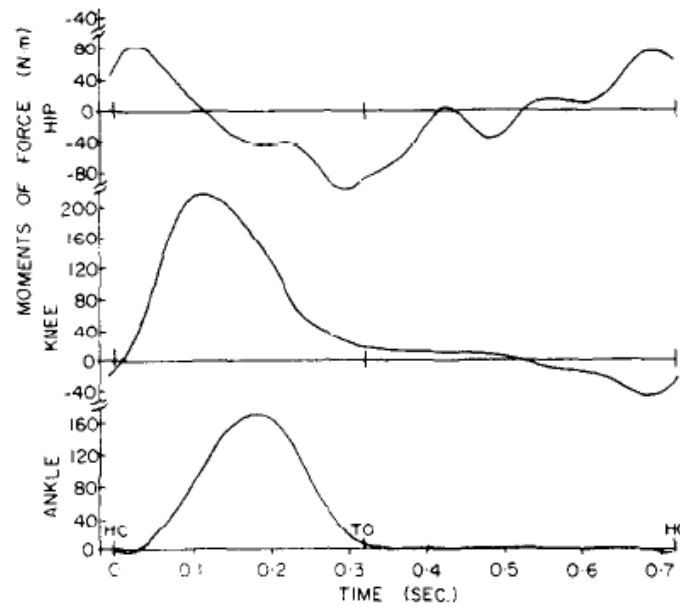


Figure 2.7. Net joint moments during a stride. Net joint moments at the hip, knee and ankle during one stride. The swing phase occurs from TO to HC. Adapted from Winter, 1983. Positive and negative moments indicate extensor and flexor moments respectively.

The exact mechanism of joint moment production depends on both contractile and non-contractile tissues, neither of which can be quantified with absolute certainty. However, the joint moments can be associated with active muscle *groups* that may contribute to those moments. During swing, the hip and knee moments are of greater magnitude and change considerably more than the ankle moment. At the hip, there is a flexor moment for about the first half of swing phase, and then an extensor moment as the foot is lowered for heel contact. At the knee, there is an extensor moment for about the first half of the phase, and then a flexor moment in the second half. Chapman and Caldwell (1983a, 1983b) reported the observed kinematic patterns and kinetic functions of the swinging lower limb in sprinting. They investigated the factors affecting energy

changes during swing, and the possible kinetic limitations due to these factors on sprint running in two international level female sprinters.

The first paper indicated that the energy generated from contralateral stance phase was greater than the TLE observed in the swinging limb (Chapman and Caldwell, 1983a). The authors proposed that energy transfer through the thigh segment and energy removal at the knee were the causes of this discrepancy. This theory is consistent with other research indicating that the two-joint muscles crossing the hip and knee determine the efficiency of energy transfer to the rest of the lower limb (Mann, Moran and Dougherty, 1986; Mero, 1985).

Chapman and Caldwell (1983b) reported results from two world-class female sprinters running at five speeds ranging from 6.71-9.49 m/s on a treadmill. This study revealed two main findings related to how the two-joint rectus and biceps femoris muscles influenced energy flow through the limb. Their first conclusion was that concentric hip flexion muscle power in late swing limits the generation and distal flow of energy through the thigh segment and to the rest of the limb (Figure 8).

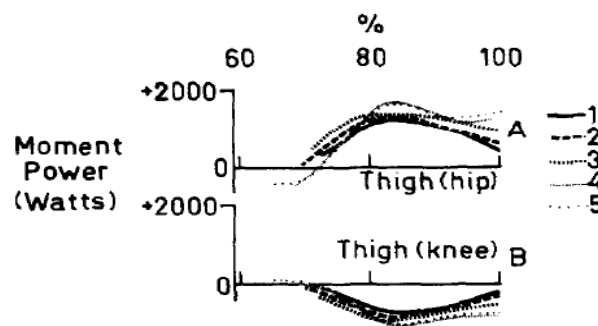


Figure 2.8. Muscle powers affecting the thigh during swing. Top graph indicates the power from the hip, and the bottom graph indicates power from the knee acting on the thigh segment across five speeds. Adapted from Chapman and Caldwell, 1983b.

The muscle moment power is positive for the thigh at the hip joint, indicating that energy is being generated to the thigh segment by muscles crossing the hip joint. The

muscle moment power for the thigh at the knee is negative, indicating energy removal from the thigh by the muscles crossing the knee joint.

A second conclusion from Chapman & Caldwell (1983b) is that eccentric knee flexion in late swing limits the amount of energy that can be removed from the lower leg (Figure 9).

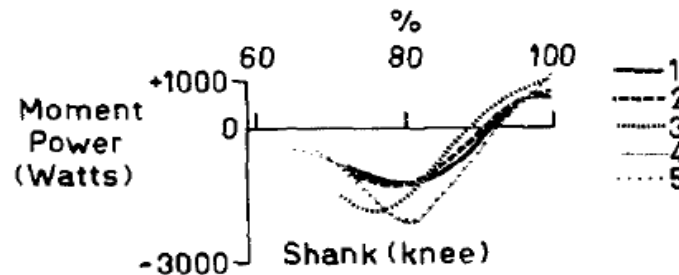


Figure 2.9. Muscle powers affecting the shank during swing. Moment power acting on the shank from the knee during a range of speeds. Chapman and Caldwell, 1983b.

No statistical difference in magnitude or timing of eccentric knee flexion muscle moment power was observed during the three fastest speeds. To increase speed further, the negative power should be larger in magnitude, and/or occur later in swing phase than the previous speed. This would allow the leg to travel faster, and the foot to be placed on the ground sooner. The fact that this does not occur implicates knee flexor muscles acting eccentrically during late swing phase as potentially failing to complete their task as effectively as possible.

Additionally, a levelling-off of peak swing limb TLE was observed as speed approached maximal, and the authors alluded to this as a mechanism potentially limiting to maximal speed (Chapman and Caldwell, 1983b). Energy generation by concentric hip extension muscle moment power during early swing could be a culprit in this inability to increase TLE further.

As a runner increases speed, they have less time to accomplish the kinetic and kinematic patterns observed during slower gait. Not surprisingly, peak net joint power, net joint muscle moment, and relative joint angular velocity values obtained for advanced sprinters were higher than those obtained for intermediate sprinters for both the hip and knee joints (Vardaxis, 1989). At maximal speed, a runner needs to generate and transfer TLE to the distal segments for rapid recovery, and delay the necessary reduction of this energy to the limb to properly position the foot for the subsequent stance phase. Therefore, it is not surprising that the biomechanical load in the hip and knee flexor muscle groups increases the most during maximal speed sprinting (Schache, 2011), and that these muscles tend to be the most frequently injured while sprinting (Croisier et al., 2002; Schache et al., 2009). These energy management issues during swing phase appear to contribute to limiting maximal sprint speed in humans.

2.9 Bilateral Analysis of Human Gait

Differences in observed kinematic patterns between limbs may limit the speed and efficiency with which the runner can move. The literature is inconsistent in this regard, with data supporting both symmetrical and asymmetrical gait patterns in walking and running (Sadeghi, Allard and Duhaime, 1997). Therefore, a bilateral investigation into differences in key kinetic parameters may aid our understanding of how increases in sprint speed are limited.

The cyclical nature of locomotion is modulated by interactions of the body's neural and mechanical dynamic systems (Masani, Kouzaki and Fukunaga, 2000). The interactions of these multiple systems result in natural variability associated with forward progression. However, excess variability implies system instability, and could be a factor

in preventing smooth gait. As a consequence, it is believed that gait variability should be kept as low as possible (Masani, Kouzaki and Fukunaga, 2000). In an opposing view, dynamical systems theory proposes that task variability is inherently functional. The degrees-of-freedom problem states that there are many methods by which humans can perform a given task, and that this redundancy may cause coordination issues during gait (Bernstein, 1967). The presence of variability in human locomotion has been well documented in terms of intra-limb (within limb) variability and inter-limb (between limbs) variability (Bartlett, 2007; Belli, 1995; Exell, 2012; Sadeghi, 1997; Vardaxis, 1989).

It has been suggested that perfect repetitions of movements are impossible due to the variations and redundancy of the neuromuscular and skeletal system (Giakas and Baltzopoulos, 1997). Because human locomotion involves numerous segmental interactions, significant variability within a subject over repeated trials is expected. Belli et al. (1995) reported variability in CoM vertical displacements of 5.2% (± 3.2) and in contact times of 1.6% (± 0.3) for healthy male track and field athletes running at 60% of their maximal speed. Bradshaw, Maulder and Keogh (2007) calculated relatively low (0.12-0.97%) inter-athlete variability values in 10 male track sprinters for three different 10m sprint times, but found variability values of roughly 50% for hip and knee angular velocities for some athletes. Zifchock and Davis (2008) used non-consecutive foot strikes to calculate between-leg differences, but also reported intra-limb variability values for knee adduction and knee joint stiffness to be 150% and 70%, respectively. Further, linear and angular velocity values are less repeatable than either temporal or positional values for both the left and right limbs regardless of SF, perhaps due to the differentiation

process used to calculate velocity from position (Karamanidis, Arampatzis and Brüggemann, 2003). Clearly, variability exists in healthy locomotion, and must be accounted for in a bilateral analysis using non-consecutive foot strikes.

In addition to intra-limb variability, there is also evidence for variation between limbs. In a study of 12 female recreational long distance runners, the intra-limb variability of hip vertical position at touchdown and toe-off and the contact times of the right body side were higher than those for the left side (Karamanidis, Arampatzis and Brüggemann, 2003). Belli et al. (1995) reported 8.9% step-to-step variability in the change in height of the CoM for 17 male subjects running at 60% maximal velocity, increasing to 13.5% as speed increased to maximal. The inherent variability from stride to stride and from step to step must be taken into account when completing studies involving non-consecutive foot strikes (Giakas and Baltzopoulos, 1997; Exell et al., 2012; Zifchock and Davis, 2008).

2.10 Asymmetry

Although intra- and inter-limb variability in human locomotion is well-established, it is important to determine if there is systematic asymmetry between legs during running. The presence of gait symmetry (SYM) and asymmetry (ASYM) has been investigated in clinical gait (e.g. Zifchock et al., 2006; Gunderson et al., 1989), able-bodied walking (e.g. Allard et al., 1996; Sadeghi et al., 1997), able-bodied running (e.g. Zifchock et al., 2008, Karamanidis, Arampatzis and Brüggemann, 2003; Vagenas and Hoshizaki, 1992), amputee sprinting (e.g. Nolan, 2008; Buckley, 1999) and able-bodied sprinting (e.g. Exell et al., 2012; Belli et al., 1995). For our purposes, symmetry will refer

to the exact replication of one limb's movement by the other while asymmetry will refer to any deviation from symmetry (Exell et al., 2012).

Researchers have developed numerical indices to quantify the degree of asymmetry for both discrete and continuous variables, although we will discuss only discrete variables in this literature review. The symmetry index (SI) was developed in an attempt to compare bilateral GRF values before and after a chiropractic manipulation (Robinson et al., 1987). The SI is a ratio of left and right discrete values shown in Equation 2.

$$SI = \left(\frac{(X_{left} - X_{right})}{0.5(X_{left} + X_{right})} \right) \times 100\% \quad (2)$$

In this equation, X_{left} and X_{right} are any discrete values for the left and right sides, and an SI value of zero indicates symmetry.

Using SI to determine asymmetry appears to be an effective method, but there is much debate as to what constitutes significant asymmetry. For example, a difference of 15% between limbs was considered clinically relevant, such that it could lead to a change in the interpretation of the data (Zifchock et al., 2008). However, SI analysis for 12 female distance runners revealed that left and right leg differences were generally less than 8% for angular displacement parameters and contact times for running at preferred SF or $\pm 10\%$ preferred SF (Karamanidis, Arampatzis and Bruggermann, 2003). Analysis of Variance (ANOVA) procedures have also been used to determine the significance of asymmetry, and are advantageous because they can discern differences between limbs not due to the nominal sample variation. Vagenas and Hoshizaki (1992) found statistically

significant levels of asymmetry for kinematic variables as low as 3% for subtalar joint angle and as high as 43% in rear foot angle for 29 male long distance runners. The different ranges for “normal” asymmetry in running are likely due to differences in calculations. The SI can be misconstrued because one has to normalize to a relatively arbitrary reference value specific to the question of interest (Zifchock et al., 2008). For example, when comparing data between injured and uninjured legs, which would serve as the reference? In addition, SI values ranging from 4 to 13,000% have been calculated for selected gait variables simply because the measured values hovered around zero, making interpretation difficult without further statistical analysis (Herzog et al., 1989). The advantages of this quick and easy SI may be countered by ambiguously chosen reference values and artificially inflated answers.

In an attempt to resolve some of the problems with SI, Zifchock et al. (2008) devised the “SYM angle” (SA), shown in Figure 10.

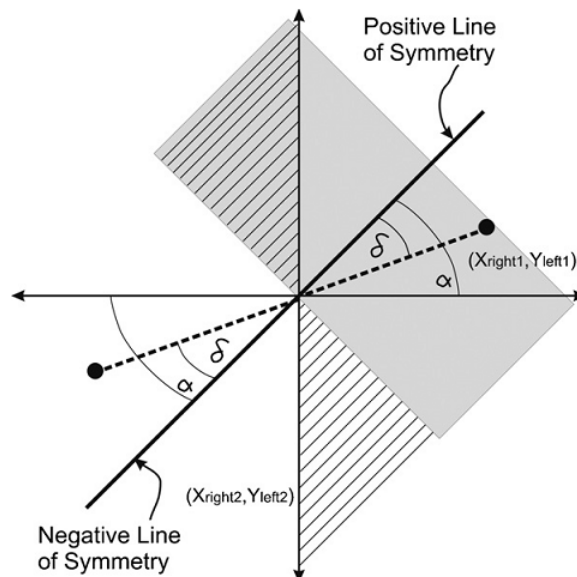


Figure 2.10. The Symmetry Angle. Pictorial representation of the SA adapted from Zifchock et al. (2008). Bold line indicates the positive and negative relationship between variables X and Y. This line bisects the horizontal at a 45° angle, and input values resulting in an SA of 45° are defined as symmetrical. Any deviation from this line (e.g. hashed line) indicates asymmetry.

The calculation of SA is outlined in Equation 3, where X_{left} and X_{right} are analogous to those variables in Equation 2.

$$SA = \frac{(45^\circ - \arctan(X_{\text{left}}/X_{\text{right}}))}{90^\circ} \times 100\%$$

But if

$$(45^\circ - \arctan(X_{\text{left}}/X_{\text{right}})) > 90^\circ$$

Then

$$SA = \frac{(45^\circ - \arctan(X_{\text{left}}/X_{\text{right}}) - 180^\circ)}{90^\circ} \times 100\%$$

(3)

This method eliminates the need to choose a reference value, since the variables may be placed in either order depending if the data fit the first or second parts of Equation 3. The SA is capped at a maximum of 100% for variables that have an equal value but are oppositely oriented, restricting the size of the asymmetries calculated and giving a more realistic representation of asymmetry present in human motion (Zifchock et al., 2008). After removing three outliers from their data set, Zifchock et al. (2008) found that SA is perfectly correlated ($r=1.00$) with SI. The decreased ambiguity and more realistic values, coupled with the correlation to SI, make SA a valuable tool to detect meaningful asymmetry in human gait. Although SA may be preferred, SI is still used by many because of its simplicity. In the following sections, asymmetry is reported using both SA and SI.

2.11 Kinematic and Kinetic Gait Asymmetry

A review of the literature on kinematic asymmetry in gait found that many assume symmetry for simplicity during data collection and analysis. But bilateral

symmetry was actually quantified in only a few studies, often with small sample sizes (Sadeghi et al., 2000). In fact, asymmetry has been widely reported for a number of kinematic variables (Gundersen et al., 1989; Vagenas and Hoshizaki, 1992; Karamanidis, Arampatzis and Bruggermann, 2003). Further some authors caution that kinematic gait parameters that provide information about the effect but not the cause of the movement might affect interpretation of lower limb behavior (Sadeghi et al., 2000).

Kinetic asymmetry was unexpectedly detected in the GRF profiles of healthy control subjects (Herzog et al., 1989), calling into question the historical assumption of bilateral gait symmetry. Asymmetries in GRF and muscle strength variables were different by only six percent between 20 injured and 20 uninjured runners (Zifchock et al., 2008). Injured runners did not consistently have more asymmetry than uninjured, agreeing with previous findings that runners with a history of tibial stress fractures exhibited similar GRF asymmetry to uninjured controls (Zifchock et al., 2006). Gait asymmetry has been documented in frontal plane GRF components of young, healthy males (Giakas and Baltzopoulos, 1997), and in EMG patterns (Ounpuu and Winter, 1989), especially for plantar flexor muscles. Furthermore, Ounpuu and Winter (1989) cautioned that pooling subject data together may conceal bilateral differences between subjects. Allard et al. (1996) reported that energy absorbed at the knee during swing was significantly less (-19.3J/kg) for the left leg than the right leg (-37.1J/kg) over 57 walking trials. Asymmetries in peak power in early swing between the right (0.94W/kg) and left (1.41W/kg) limbs were determined in healthy older adults (Sadeghi et al., 2004). Thus, there is ample evidence for kinetic asymmetry in both stance and swing, and these are likely linked to kinematic asymmetry.

2.12 Asymmetry and Maximal Velocity Running

There is sparse literature describing kinetic asymmetry and its relationship to maximal velocity running. Using a kinematic arm measurement tool, Belli et al. (1995) calculated variability and asymmetry of contact time and CoM displacement in 6116 individual steps of 19 male long distance runners. The authors found that the relative stride to stride variability of both vertical CoM displacement and contact time did not change as running speed increased from 60% to 100% maximal aerobic velocity (as determined by an incremental VO_2 max test). The participants then sprinted at their actual maximal speed on the treadmill and vertical CoM displacement variability increased from $7.5 \pm 1.8\%$ at maximal aerobic velocity to $12.0 \pm 3.0\%$, at actual maximal speed. Contact time variability at maximal aerobic velocity ($1.8 \pm 0.5\%$) increased to $3.6 \pm 1.7\%$ at actual maximal speed. The authors used step and stride variability to indicate bilateral variance within and between strides respectively. Relative step asymmetry was calculated as the difference between step variability and stride variability. Therefore, asymmetry was mathematically defined here as the difference between limbs not attributable to the inherent differences between strides. ASYM's contribution to step variability was calculated by dividing relative step asymmetry by step variability and multiplying by 100% (Belli et al., 1995). ASYM's contribution to step variability rose as speed increased, but so did stride variability, and asymmetry could never explain more than 50% of the differences seen from stride to stride. So asymmetry was present, but was classified as insignificant due to variability. Asymmetry cannot be inferred without a measure of stride variability, even for analyses of consecutive running trials (Belli et al., 1995).

The concept of inherent stride variability discussed earlier was a main component of two recent papers by Exell et al. (2012a, 2012b) concerning a group of 8 male sprinters. In the first study, the authors used a combination of variability and SA measures to determine the degree of asymmetry in runners sprinting at maximal speed. Because the authors recorded kinetics from non-consecutive steps, asymmetry was deemed significant only if the bilateral difference was larger than the intra-limb variability between trials. There were a wide range of SA values (3% to 77%) for which significance was determined, indicating drastically different levels of variability between kinematic and kinetic parameters during maximal velocity sprinting (Exell et al., 2012a). Every subject's kinetic asymmetry values were larger than their corresponding kinematic values, yet only 23% of kinetic variables displayed significant asymmetry compared with 39% of kinematic variables (Exell et al., 2012a). The authors concluded that there is more variability in kinetic than kinematic parameters at maximal velocity, conjecturing that the neuromuscular system may be kinetically compensating in an attempt to minimize the kinematic variability. No two participants displayed significant asymmetry for the same combination of kinetic and kinematic variables, indicating the importance of a subject-specific approach to asymmetry. The authors further suggested that asymmetry for a given gait parameter is meaningful only if its inter-limb exceeds its intra-limb variability.

The second study's aim was to combine kinematic and kinetic asymmetry scores to quantify athlete asymmetry during sprint running (Exell et al., 2012b). SA scores for set of pre-determined variables deemed important by coaches were used to construct composite kinematic and kinetic asymmetry scores. A variable was only included in these scores if its inter-limb differences were significantly larger than those within-limbs. As in

the previous study, kinematic SAs were smaller, ranging from ~0.6% to 6.7%, while the kinetic SAs were larger (3% to 45%), so the kinetic composite asymmetry scores were larger than the kinematic composite scores for every athlete tested. Although a single asymmetry score may seem useful as a gross indicator, it paints an incomplete picture of asymmetry for an individual athlete. There was no relationship between the athletes with the highest kinematic and kinetic asymmetry scores, in fact two athletes with similar kinetic asymmetry scores had both the highest and the lowest kinematic asymmetry scores (Exell, 2010). The authors suggest that the composite scores are merely precursors for further analyses, and that examination on an athlete-by-athlete basis is needed to determine the effects of asymmetry on specific performances (Exell et al., 2012b). Specific kinetic causes for a particular kinematic asymmetry can never be deduced with a single-score index. Furthermore, there was no correlation between sprint performance and the magnitude of the composite score (Exell et al., 2012b). Finally, these SAs and composite scores were determined at maximal speed only, giving no indication of how asymmetry changes as speed progresses to maximal, or whether asymmetry might play a role in limiting the maximal speed. Overall, information on asymmetry and its relationship to maximal running velocity is currently lacking.

2.13 Variability and Asymmetry during Athletic Performance

Both variability and asymmetry have been linked with a decrease in athletic performance in humans (Vagenas and Hoshizaki, 1992; Bradshaw, Maulder and Keogh, 2007; Flanagan and Harrison, 2007; Masani, Kouzaki and Fukunaga, 2002; Ball and Scurr, 2011), as well as bi- and quadra-pedal animals (Martin, 2001; Garland and Freeman, 2005). Hind limb length asymmetry is significantly less in mice bred for high

endurance running compared to controls (Garland and Freeman, 2005). A study of 29 male lizards revealed that the lizard with the most symmetrical femur lengths had a maximal escape speed 30% faster than the lizard with the least symmetrical femur lengths (Martin and Lopez, 2001). To offset anatomical asymmetries such as leg length discrepancies, a human runner's neuromuscular system will be forced to perform compensatory adjustments that present as impaired running technique (Vagenas and Hoshizaki, 1992).

Perhaps the best example of anatomical asymmetry is unilateral amputees, who display well-documented asymmetrical gait with respect to joint and segment kinematics and kinetics (Nolan, 2008). Much interest has centered on the Paralympic and Olympic Games, and the possible crossover between them. Recently, a bilateral amputee has achieved split speeds during 400m sprint events competitive with the best able-bodied athletes in the world (Bruggeman et al., 2008). Although support exists both for (Weyand et al., 2009) and against (Kram et al., 2010) the possibility that an amputee sprinter may have an advantage over able-bodied athletes, there is no doubt that bilateral amputees have some advantage over unilateral amputees. The fastest double transtibial amputee athletes are able to perform in the 200- and 400-m sprint events faster than the fastest single leg amputee, possibly due to the asymmetric nature of the prosthetic and the healthy limbs of the unilateral amputee (Bruggeman et al., 2008). For example, during swing phase the total work done by the prosthetic limb of a unilateral amputee runner is similar to that of an able-bodied runner, while the sound limb exhibits a 69% increase in the total work done (Buckley, 1999).

In contrast, there is evidence for variability and asymmetry occurring in subjects with no obvious lower limb anatomical discrepancies. A study of starting block kinematics of 10 male national- and regional-standard sprinters revealed that the fastest sprinter had only 1.75% variation in the length of his second stride; about 25 times less variation than the slowest sprinter (Bradshaw, Maulder and Keogh, 2007). Masani, Kouzaki and Fukunaga (2002) indicate that a healthy neuromuscular system suppresses gait variability well, whereas gait variability rises with neuromuscular system impairment, as in gait-disabled patients. Leg preference laterality can result in a bilaterally asymmetric response in force or neuromuscular variables, and is associated with poor performance during demanding mechanical and neural exercise (Ball and Schurr, 2011). Asymmetry was also found in 8 healthy adult males completing a drop jump task in which they were asked to jump as high and as fast as possible after falling from a height. Both the flight times and “reactive strength” indices were significantly higher in the subjects’ dominant legs compared to their non-dominant (Flanagan and Harrison, 2007). The authors stated that bilateral stiffness imbalances could be detrimental to performance or could increase soft tissue injury risk. In contrast, a bivariate linear regression analysis revealed that SI had no relation to sprint speed, leading the authors to conclude that asymmetry is not related to maximal velocity attainment (Korhonen, et al., 2010). However the limitations of using SI have been previously discussed, and may have affected the outcome of their regression analysis. Swanson and Caldwell (2005) found unique bilateral differences between peak knee and hip powers implicated in energy absorption and generation respectively at maximal speed in 5 of their 8 subjects. The authors suggested that bilateral asymmetry and increased

variability in hip and knee kinetics at maximal speed compared to slower speeds may indicate a failure in lower extremity coordination during swing that hinders effective force application during the subsequent stance phase.

2.14 Conclusion

The literature points to many different potential limitations to maximal running velocity. Kinetic factors describing the causes of kinematic motion are advantageous to investigate because the mechanisms behind sprint running can be elucidated. SF increases to achieve higher velocity, and moving faster through the gait cycle has implications for the kinetics during both stance and swing phases. Variability and asymmetry in kinetic measures between legs has been linked with decreased performance, although there is currently no definitive link between asymmetry and maximal velocity limitation.

CHAPTER 3

METHODS

The experimental protocol of this study involved the collection of kinematic and kinetic data during four sprinting conditions. The data provided measures of key kinematic and kinetic parameters associated with high speed running. Asymmetry of these key variables was examined as the speed of the runner increased from submaximal to maximal.

3.1 Subjects

The subjects were 9 female college-aged (21 ± 3 years) speed and power athletes (height = 1.64 ± 0.07 m, mass = 60.58 ± 7.48 kg). These subjects were experienced soccer (n=4), lacrosse (n=1), sprint (n=1), rowing (n=1), martial arts (n=1) and rugby (n=1) athletes currently competing at the collegiate level. All subjects had no lower limb musculoskeletal injuries during the previous 6 months, and no injuries requiring surgery within the previous year. These conditions were an attempt to decrease the variance inherent in the sprinting motions and maximal speeds. An a priori sample-size estimation procedure using expected differences and means from previous literature (Exell 2010; Belli et al, 1995) indicated that 9 subjects were needed to achieve significance for bidirectional independent samples ($\alpha=0.05$, $\beta=0.20$). All of the subjects signed an informed consent and modified health history information documents (Appendix D).

3.2 Protocol

Each subject took two sessions to complete the study separated by no more than 10 days to avoid a potential training effects. The first session allowed the subjects to familiarize themselves to the force treadmill. Also, during this session the subject's

maximal speed was determined by having the subject complete 5-6 trials of 4 second sprint bouts on the force treadmill. Maximal speed was reached when the subject, upon visual inspection, could not maintain a constant position on the treadmill belt as described by Thelen and colleagues (2005). These bouts were separated by rest intervals during which the subject's heart rate was lowered to approximately 100 beats per minute (Thelen et al, 2005). No kinematic or kinetic data was collected during this session, although anthropometric parameters were recorded.

During the subject's second session, retro-reflective markers were placed bilaterally on their toes, 5th metatarsal heads, 1st metatarsal heads, lateral malleoli, medial malleoli, lateral femoral condyles, medial femoral condyles and greater trochanters. In addition to the limbs, retro-reflective markers were also placed bilaterally on the ASIS, PSIS, and the sacrum to outline the pelvis segment (Figure 3.1).

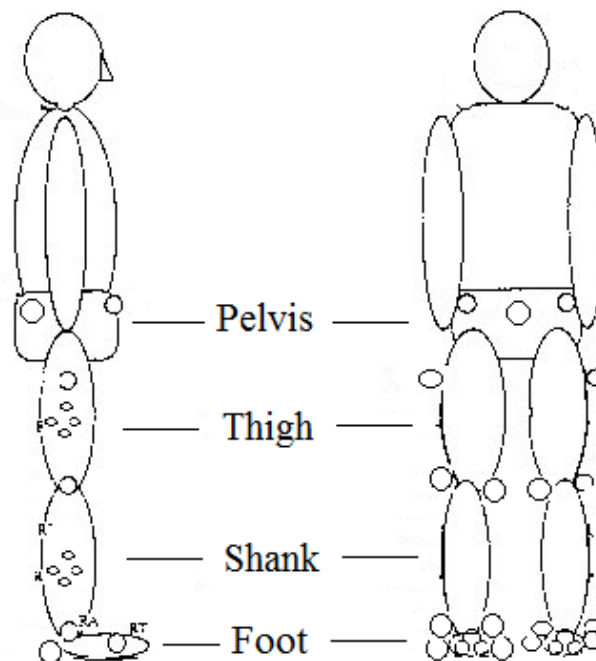


Figure 3.1. Data Collection marker Set. Marker set up for the data collection trials. Markers are indicated by (o) attached to labelled pelvis, thigh, shank and foot segments. Adapted from Hamill and Selbie (2004).

Plastic plates of 4 non-collinear tracking markers were secured to the thigh and shank segments, while plates of 3 non-collinear tracking markers were placed on both heels. These plates were used during the sprinting trials so that some calibration markers (Figure 3.1) that might interfere with the natural sprinting motion could be removed. Positional data from these markers were collected three-dimensionally throughout all trials.

Each subject performed a stationary standing calibration trial on the level force treadmill to relate the positions of the markers to specific limb segments and to the force transducers of the treadmill (Hamill and Selbie, 2004). Previous research indicates that 8-10 footfalls per limb are required to detect meaningful differences in GRF profiles between conditions for the same subject (Bates, Osternig and Sawhill, 1983). Therefore, the standing calibration trial was followed by a series of eight sprint trials of five seconds each on the level treadmill completed in a “speed pyramid” outlined in Table 3.1.

Table 3.1. Chronological order of Sprint Conditions. Percentages were determined from the Maximum speed reached during the subject’s initial visit.

Trial	1	2	3	4	5	6	7	8
Condition (% Max)	70	80	90	100	100	90	80	70

This fixed order was determined a priori because previous research indicated that completely randomizing the trial order during high speed running could lead to problems of feasibility, ethicality and accuracy by having a subject attempt a maximal speed trial first (Hamill et al., 1983; Thelen et al., 2005). This “speed pyramid” was also employed to ensure a sufficient number of steps were collected while minimizing the chance of injury and fatigue. As in the previous visits, subjects wore the safety harness during all trials. Kinematic and kinetic data (see below) were collected during each trial. Subjects

again had the opportunity for active and passive recovery between trials until their heart rate recovered to below 100 bpm as per recommendations from Thelen et al. (2005), calculated by the carotid palpation method. If the subject either exceeded or failed to reach the desired top speed, the speed of the treadmill was increased or decreased by 0.2 m/s respectively. Subject heart rates were returned to 100 bpm, and the trial was repeated at the new speed. Submaximal conditions remained the same so as not to fatigue the subjects.

3.3 Data Collection

Three-dimensional kinematic data from the retro-reflective markers was collected with a set of 8 Pro-Reflex Cameras coordinated by Qualysis Track Manager (QTM; Qualysis, Gothenburg, Sweden) at a sampling rate of 240 Hz. Kinetic GRF data were collected from four treadmill force transducers (Advanced Medical Technology Inc., Watertown, Ma.) at a sampling rate of 1200 Hz, with kinematic and force data synchronized within QTM.

3.4 General Data Analysis

The raw data input to QTM during the collection were exported to the biomechanical modeling software Visual 3-D (C-Motion Inc, Germantown, MD). In Visual 3-D, a linked-segment model was generated from the calibration trial for further analysis of the data using previously validated 3-dimensional regression equations to determine segment length, mass and inertial parameters (Hanavan,1964). An X, Y, Z Cardan rotation sequence was used to define the rotational motion of the modeled lower limb segments.

Marker position data were low pass, bi-directionally filtered using a 4th order Butterworth filter with a cut-off frequency of 6Hz. An inverse kinematics model was used to compute the position and orientation of the segments during the sprinting trials. This model was used instead of the 6 degrees-of-freedom model because there were some instances of marker dropout that negatively affected tracking segmental motion, especially of the foot during mid-swing when it is closest to the pelvis. The center of mass acceleration of feet and shank segments were used to determine instances of foot strike and toe off using algorithms detailed by Hreljac and Stergiou (2000). This procedure was used because the force and center of pressure (COP) signals exported from the force treadmill were too noisy to provide a faithful determination of these key gait events, while kinematic gait event determination has been validated for both walking (Hreljac and Marshall, 2000) and running (Hreljac and Stergiou, 2000).

Kinetic force data were first bias-removed using a flight phase baseline software algorithm, then low pass, bi-directionally filtered using a 4th order Butterworth filter with a cut-off frequency of 20 Hz. The filtered treadmill force data were used to calculate the center of pressure (CoP) on the treadmill and on the runner's foot (Hamill and Selbie, 2004). The CoP data were low pass, bi-directionally filtered using a 4th order Butterworth filter with a cut-off frequency of 20 Hz, and then cleaned by eliminating any incorrectly computed CoP values during identified flight phases. The kinematic, CoP and GRF data were to be used to determine lower extremity net joint moments and joint reaction forces using an inverse dynamics analysis (Robertson, 2004). There were errors associated with the CoP signal in stance near foot strike and toe off, when the vertical forces are small and the calculated CoP location is largely dictated by the relatively noisy A/P transducer

forces. To minimize these errors, the CoP signal was bounded to lie between the toe and heel marker positions during stance. If the A/P component of the CoP signal fell outside of the bounds of the foot, it was replaced with the A/P position of the CoM of the foot at that frame. A cross-validatorspline smoothing and differentiation procedure (Woltring, 1986) was used to smooth and interpolate the CoP data between foot strikes and toe offs that defined the stance phases in an effort to minimize these CoP replacement errors. Overall, any residual CoP errors after this process were not of concern because the GRF forces near foot strike and toe off were very low.

3.5 Temporal Analysis

Stride and step frequencies were assessed using the number of contralateral and ipsilateral foot contacts in one second respectively. Stride and step lengths were calculated as the product of treadmill belt speed and time between consecutive contralateral and ipsilateral foot strikes, respectively. Stride and step velocities were calculated as the product of the frequencies and lengths of strides and steps respectively. Stance time and swing time are the length of time spanning ipsilateral foot strike to toe off and toe off to foot strike respectively.

3.6 Kinetic Analysis

Horizontal and vertical impulses were calculated during both the propulsive and braking phases of stance as designated by the horizontal GRF direction. Impulse was calculated as the integral of horizontal GRF during the time of force application ($I_H = \int GRF_H dt$) during each phase. Vertical propulsive and braking impulse was calculated as the integral of vertical GRF during the time of force application ($I_V = \int GRF_V dt$) during each phase. Effective vertical stiffness (EK_{vert}) was calculated as the ratio of the peak

vertical GRF and the maximum vertical displacement of the sacral marker (representing the total body center of mass (CoM)) during the stance phase. Muscle moment power, and limb energetics were calculated for both the stance and swing phases using the equations from Chapman and Caldwell (1983a), as described in Appendix B. The occurrence of the peak muscle powers were normalized to 100% of phase time.

3.7 Symmetry

Symmetry (or Asymmetry) was assessed using a modified Symmetry Angle equation (AbsSA) from Zifchock et al. (2008; Equation 3.1).

$$\text{AbsSA} = \frac{\left| 45^\circ - \arctan(X_{\text{left}}/X_{\text{right}}) \right|}{90^\circ} \times 100\% \quad (4)$$

X_{left} and X_{right} refer to the left and right values of any temporal or kinetic parameter of interest. The following example illustrates the effectiveness of this asymmetry index.

3.7.1 Absolute Asymmetry Index Example

Take a hypothetical task for which the left and right limb values to be compared are presented in Table 3.2. Ratios are formed for set A and B using first the left and then the right values as the numerator.

Table 3.2. Hypothetical Data Set. Left and right limb values and quotients.

Set	A	B
Left	10	10
Right	5	15
Left / Right	2	0.67
Right / Left	0.5	1.5

These ratios and their corresponding absolute asymmetry scores can be located by the dashed (Set A) and solid (Set B) vertical lines on Figure 3.2. The same absolute

asymmetry score can result from two different ratios, so it is a robust measure of asymmetry because it is insensitive to the choice of which limbs' value is the numerator. Consequently, differences of 50% (quotients of 0.5 and 2) and ~33% (quotients of 0.667 and 1.5) between limbs will always yield absolute asymmetry scores of ~20 and ~13 respectively regardless if the numerator is larger or smaller than the denominator. In this way, it eliminates the need for the researcher to arbitrarily choose a numerator and denominator.

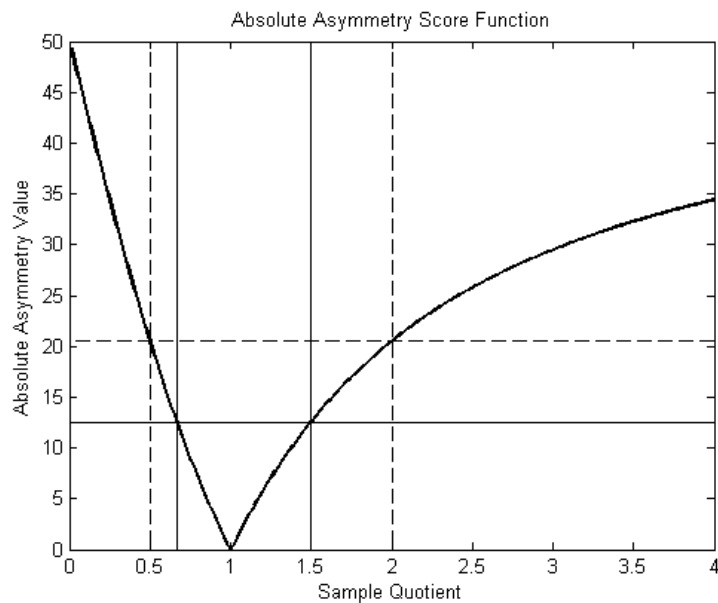


Figure 3.2. Absolute Asymmetry Score Function. Absolute Asymmetry Scores (AbSAs) for varying ratios (thick black line). The intersection of dashed lines represents the AbSA values corresponding to Set A. The intersection of thin, solid lines represents the AbSA values corresponding to Set B.

3.8 Kinematic and Kinetic Asymmetry

Although kinematic asymmetry is not the primary focus of this study, asymmetries between bilateral linear and angular positions, velocities and accelerations of the segments and joints were used to aid in the interpretation of kinetic asymmetries. The specific parameters that were assessed for bilateral asymmetry are listed in Table 3.3.

Table 3.3. Key Asymmetry Parameters.

Stride Characteristic	Parameter
Temporal/ Descriptive	Stride Length, Step Length, Step Time, Flight Time, Stance Time, Peak Eccentric Knee Power Timing
Stance Phase	Vertical Propulsive Impulse, Horizontal Braking Impulse, Horizontal Propulsive Impulse, Effective Vertical Stiffness
Swing Phase	Peak Concentric Hip Extension Muscle Power, Peak Eccentric Knee Flexion Muscle Power

Custom MATLAB™ (R2010a, The MathWorks, Natick, MA, USA) software was written which calculated absolute asymmetry scores for the parameters listed in Table 3.3. These scores were organized by parameter and exported to Microsoft Excel© (2007, Redmond, WA, USA) software.

3.9 Statistical Analysis

To ensure traditional statistical procedures could be used to examine our data, all asymmetry data were tested for normality. The differences between the mean and median for all parameters were less than or equal to 10%, so the asymmetry data were considered normally distributed (Exell, 2012b) and traditional analysis of variance (ANOVA) tests were implemented. For each parameter, a three-factor (speed, trial, subject) repeated-measures ANOVA (SAS version 9.1.3, SAS Institute Inc., Cary, NC, USA) was used to examine how absolute asymmetry scores differed across speed. In addition, the interaction of speed and trial order was examined so as to elucidate potential order effects on asymmetry. Tukey's post-hoc mean separation and linear contrast procedures were applied to examine any significant interactions or trends respectively.

CHAPTER 4

RESULTS

4.1 General Speed and Temporal Characteristics

The general speed characteristics (Mean \pm SD) for the 9 sprinters across the 4 speed conditions are presented in Table 4.1. Maximal speeds ranged from 7 to 8.2 m/s, and temporal gait characteristics are presented in Table 4.2.

Table 4.1. General Speed Characteristics. Average absolute and percent max speed during the four speed conditions (70, 80, 90, 100% Max speed). Here and in other tables, the speed conditions will be represented with respect to the percentage of maximal speed (i.e. “70” is the 70% maximal speed condition).

	70	80	90	100
Average Speed (m/s)	5.43 (\pm 0.29)	6.20 (\pm 0.32)	6.98 (\pm 0.36)	7.72 (\pm 0.34)
Average Speed (%)	70.29 (\pm 1.45)	80.33 (\pm 1.65)	90.37 (\pm 1.86)	100 (\pm 0)

Table 4.2. Temporal Gait Characteristics. Mean \pm (SD) across speed conditions for both the left (L) and right (R) limbs.

Temporal Gait Characteristics	70		80		90		100	
	L	R	L	R	L	R	L	R
Stride Frequency (strides/s)	1.66 (0.12)	1.66 (0.12)	1.77 (0.12)	1.77 (0.13)	1.91 (0.14)	1.91 (0.13)	2.08 (0.19)	2.05 (0.13)
Stride Length (m)	3.27 (0.14)	3.27 (0.14)	3.52 (0.14)	3.52 (0.14)	3.67 (0.15)	3.66 (0.12)	3.76 (0.18)	3.77 (0.14)
Stride Time (s)	0.60 (0.03)	0.60 (0.05)	0.57 (0.03)	0.57 (0.04)	0.53 (0.02)	0.53 (0.04)	0.49 (0.02)	0.49 (0.03)
Step Frequency (steps/s)	2.48 (0.17)	2.47 (0.19)	2.60 (0.19)	2.61 (0.20)	2.82 (0.17)	2.82 (0.19)	3.01 (0.18)	3.03 (0.18)
Step Length (m)	2.19 (0.13)	2.20 (0.11)	2.39 (0.10)	2.38 (0.11)	2.48 (0.09)	2.48 (0.10)	2.57 (0.08)	2.56 (0.08)
Step Time (s)	0.30 (0.02)	0.30 (0.01)	0.28 (0.01)	0.29 (0.01)	0.26 (0.01)	0.26 (0.01)	0.24 (0.01)	0.25 (0.01)
Swing Time (s)	0.41 (0.03)	0.41 (0.03)	0.39 (0.03)	0.39 (0.03)	0.36 (0.02)	0.36 (0.03)	0.33 (0.02)	0.33 (0.02)
Stance Time (s)	0.20 (0.03)	0.20 (0.02)	0.18 (0.02)	0.18 (0.01)	0.17 (0.02)	0.17 (0.01)	0.16 (0.01)	0.16 (0.01)
Flight Time (s)	0.10 (0.02)	0.11 (0.01)	0.10 (0.01)	0.10 (0.01)	0.09 (0.01)	0.09 (0.01)	0.09 (0.01)	0.09 (0.01)

With speed increases, stride and step frequencies both increased by ~20%, while the stride and step lengths increased ~14% from the slowest to fastest speeds. Stride, step, swing, stance and flight times all decreased with speed; reaching a minimum at top speed. Stance and swing times decreased comparatively more than flight times (~20% to ~10% respectively). Variability as indicated by standard deviations was also similar across speeds for these temporal parameters. The general strategy by which subjects increased speed can be visualized in Figure 4.1. Subjects increased both stride length and frequency to attain higher speeds, with stride frequency increasing more sharply as they approached the fastest speed.

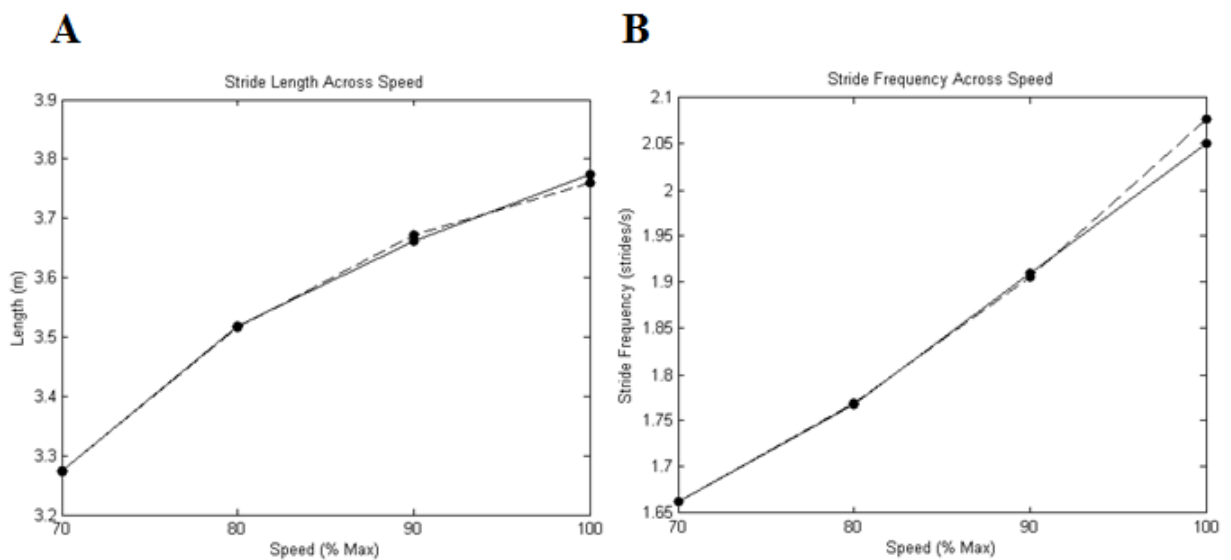


Figure 4.1. Stride Length and Frequency Across Speed. Left (dashed) and right leg (solid) stride length (A) and frequency (B) changes with increasing speed.

4.2 Kinetics and Kinematics

General gait kinetic information describing how subjects were able to increase stride length and frequency is displayed in Table 4.3. Greater peak vertical ground reaction forces (GRFs) were generated by the subjects as speed increased, with top speed vertical GRFs ~ 12% larger than those at the slowest speed. However, vertical propulsive

impulses were ~ 40% less at maximal speed compared to the 70% speed condition, due to the decreased contact times. A/P GRF propulsive and braking impulses both decreased from the slowest to fastest speeds, more so for propulsive impulses (~ 37% decline) than braking impulses (~ 23% decline).

Table 4.3. General Gait Kinetics. (Mean \pm SD) during left (L) and right (R) foot contacts across the four speed conditions (% Max).

Gait Kinetics Mean (\pm SD)	70		80		90		100	
	L	R	L	R	L	R	L	R
Peak Vertical GRF (N)	1664.40 (192.08)	1691.70 (197.99)	1723.90 (181.41)	1764.50 (180.20)	1811.70 (175.85)	1807.30 (172.14)	1907.00 (224.86)	1907.00 (182.89)
A/P Brake Impulse (N*s)	-14.36 (1.34)	-14.66 (2.74)	-13.91 (1.56)	-14.72 (2.17)	-13.59 (1.88)	-13.53 (2.14)	-13.37 (2.81)	-13.9 (2.05)
A/P Propulsive Impulse (N*s)	14.12 (1.53)	14.31 (1.73)	13.43 (1.76)	13.99 (1.45)	13.09 (2.03)	13.31 (1.58)	12.09 (2.03)	12.48 (1.75)
Vertical Brake Impulse (N*s)	119.07 (15.46)	121.21 (19.57)	113.64 (16.91)	116.59 (17.72)	108.29 (14.69)	107.32 (14.78)	101.23 (19.47)	102.43 (15.2)
Vertical Propulsive Impulse (N*s)	69.62 (8.25)	71.69 (8.68)	64.13 (8.24)	66.4 (8.68)	59.4 (7.3)	61.13 (7.18)	55.87 (5.65)	55.71 (7.24)

Tables 4.4 and 4.5 describe metrics drawn from joint kinematics and kinetics, respectively, across speeds. Example time series plots for these kinematic and kinetic variables are presented in Appendix H.

Table 4.4. Joint Kinematics. Left (L) and right (R) ankle, knee and hip joint kinematics (Mean \pm SD) across speed conditions.

Joint Kinematics	70		80		90		100	
	L	R	L	R	L	R	L	R
Peak Dorsi-Flexion Angle (deg)	19.87 (13.26)	22.28 (5.15)	21.51 (3.57)	20.87 (5.42)	16.56 (13.2)	18.67 (5.18)	18.26 (4.94)	16.27 (6.26)
Peak Plantar-Flexion Angle (deg)	-29.48 (4.9)	-28.08 (7.72)	-32.13 (4.78)	-31.48 (6.27)	-34.94 (5.23)	-33.85 (7.64)	-37.34 (4.37)	-37.01 (5.92)
Ankle ROM (deg)	49.35 (8.36)	50.36 (2.56)	53.64 (1.21)	52.35 (0.85)	51.5 (7.97)	52.52 (2.46)	55.6 (0.57)	53.28 (0.35)
Peak Dorsi-Flexion Velocity (deg/s)	336.63 (71.68)	316.98 (50.06)	364.85 (93.32)	345.68 (87.94)	341.38 (102.12)	336.49 (71.17)	323.96 (77.16)	302.33 (96.04)
Peak Plantar-Flexion Velocity (deg/s)	-699.74 (72.2)	-693.73 (84.77)	-741.25 (58.12)	-730.44 (68.3)	-756.63 (89.12)	-744.09 (51.7)	-775.69 (45.05)	-750.64 (62.14)
Min Knee Flexion Angle (deg)	-9.78 (7.98)	-12.33 (5.19)	-11.48 (5.83)	-13.65 (5.53)	-14.59 (9.7)	-15.22 (10.74)	-18.93 (7.46)	-19.55 (8.32)
Peak Knee Flexion Angle (deg)	-131.51 (5.44)	-130.15 (8.07)	-136.69 (4.68)	-134.35 (7.61)	-139.53 (5.74)	-136.94 (7.54)	-140.45 (5.18)	-139.59 (7.09)
Knee ROM (deg)	121.73 (2.54)	117.82 (2.88)	125.21 (1.15)	120.7 (2.08)	124.94 (3.96)	121.72 (3.2)	121.51 (2.28)	120.04 (1.24)
Peak Knee Extension Velocity (deg/s)	974.22 (84.99)	947.49 (78.55)	1039.8 (94.17)	1012.3 (88.66)	1095.1 (81.75)	1082.6 (126.39)	1086.1 (73.86)	1084.7 (58.71)
Peak Knee Flexion Velocity (deg/s)	-924.32 (64.51)	-889.71 (72.22)	-992.01 (85.39)	-959.22 (89.09)	-1087.9 (89.83)	-1034.4 (87.78)	-1136.6 (70.4)	-1113.7 (66.69)
Peak Hip Flexion Angle (deg)	62.03 (9.93)	60.89 (11.56)	66.32 (9.66)	65.66 (10.23)	70.09 (9.68)	65.81 (17.48)	72.18 (6.22)	72.94 (6.42)
Peak Hip Extension Angle (deg)	-24.64 (9.23)	-22.28 (7.02)	-25.21 (7.48)	-23.52 (7.57)	-28.24 (14.03)	-22.09 (8.86)	-24.9 (7.4)	-23.53 (7.33)
Hip ROM (deg)	86.66 (0.7)	83.17 (4.54)	91.53 (2.18)	89.18 (2.66)	98.33 (4.35)	87.9 (8.62)	97.08 (1.18)	96.47 (0.9)
Peak Hip Flexion Velocity (deg/s)	567.7 (53.53)	548.84 (81.91)	628.55 (54.29)	615.17 (75.81)	667.53 (64.84)	613.77 (173.6)	700.32 (49.53)	685.96 (87)
Peak Hip Extension Velocity (deg/s)	-581.09 (56.63)	-574.01 (53.94)	-646.11 (73.54)	-633.27 (74.27)	-694.47 (158.89)	-626.72 (159.3)	-717.22 (65.2)	-681.68 (95.43)

Table 4.5. Joint Kinetics. Left (L) and right (R) peak joint moment, power and segmental energy (Mean \pm SD) across speed.

Joint Kinetics	70		80		90		100	
	L	R	L	R	L	R	L	R
Peak Dorsi-Flexion Moment (Nm)	4.23 (1.12)	4.29 (1.2)	4.99 (1.42)	4.99 (1.38)	5.65 (1.4)	5.71 (1.68)	6.13 (1.32)	6.24 (1.66)
Peak Plantar Flexion Moment (Nm)	-104.14 (34.7)	-109.36 (40.2)	-111.51 (33.19)	-119.37 (43.55)	-110.75 (28.03)	-115.43 (32.8)	-119.25 (28.99)	-119.21 (35.62)
Max Ankle Power (W)	718.49 (235.91)	735.86 (199.56)	806.34 (165.19)	836.41 (291.83)	816.57 (228.87)	818.8 (254.79)	831.2 (188.14)	834.17 (233.38)
Min Ankle Power (W)	-294.94 (164.05)	-305.57 (185.44)	-368.75 (198.53)	-372.83 (221.96)	-371.87 (198.45)	-352.18 (165.81)	-403.16 (239.22)	-383.97 (237.11)
Peak Knee Extension Moment (Nm)	231.21 (47.73)	241.07 (45.61)	229.23 (31.1)	231.05 (40.97)	208.05 (44.34)	208.37 (37.94)	194.9 (39.88)	194.26 (41.19)
Peak Knee Flexion Moment (Nm)	-57.02 (12.68)	-56.39 (12.88)	-68.08 (15.13)	-67.61 (15.23)	-85.47 (17.93)	-83.16 (17.56)	-96.96 (19)	-97.37 (18.22)
Max Knee Power (W)	898.73 (326.74)	931.77 (259.86)	928.11 (251.89)	908.07 (298.18)	747.82 (284.21)	758.03 (305.89)	659.89 (199.79)	643.32 (307.37)
Min Knee Power (W)	-1153.2 (344.37)	-1172.3 (314.57)	-1142.8 (294.99)	-1138.4 (352.65)	-1478.4 (304.81)	-1357 (304.82)	-1589.6 (274.51)	-1620.8 (308.61)
Min Knee Power Timing (% Swing)	80.37 (1.64)	79.96 (1.53)	79.92 (0.9)	79.67 (2.1)	79.19 (3.01)	80.04 (1.18)	80.4 (2.01)	80.13 (2.03)
Peak Hip Flexion Moment (Nm)	109.82 (30.53)	114.69 (30.48)	132.78 (35.57)	137.13 (37.08)	159.6 (43.03)	164.68 (42.16)	183.12 (39.52)	190.35 (44.79)
Peak Hip Extension Moment (Nm)	-155.09 (49.48)	-151.57 (45.11)	-154.92 (57.32)	-157.39 (58.64)	-171.04 (39.45)	-168.89 (41.5)	-199.06 (45.97)	-199.66 (42.67)
Max Hip Power (W)	554.37 (128.02)	538.35 (128.13)	718.92 (166.32)	671.98 (211.53)	1034.4 (224.92)	933.78 (243.83)	1274.8 (240.89)	1240.5 (294.34)
Min Hip Power (W)	-610.46 (227.07)	-605.27 (309.67)	-676.43 (192.18)	-620.37 (259.64)	-868.37 (304.13)	-800.07 (279.68)	-1062.1 (365.09)	-1020.6 (348.59)
Max Foot Energy (J)	18.96 (4.6)	19.08 (4.19)	23.98 (5.71)	23.91 (4.99)	29.46 (6.28)	28.48 (5.89)	32.37 (5.5)	32.69 (5.81)
Min Foot Energy (J)	1.5 (0.19)	1.61 (0.37)	1.75 (0.48)	1.72 (0.32)	1.99 (0.51)	1.98 (0.41)	3.11 (3.51)	2.29 (0.51)
Max Shank Energy (J)	33.29 (6.25)	32.68 (5.63)	39.4 (7.09)	39.06 (6.9)	45.88 (9)	44.82 (8.21)	52.06 (9.06)	51.44 (9.23)
Min Shank Energy (J)	11.59 (1.32)	11.56 (1.63)	12.74 (1.68)	12.5 (1.69)	14.23 (2.31)	13.9 (1.88)	16.17 (3.98)	15.56 (2.38)
Max Thigh Energy (J)	51.03 (7.36)	50.4 (7.76)	54.81 (8.23)	53.41 (7.73)	58.31 (9.19)	57.05 (8.09)	59.96 (8.41)	59.51 (9.32)
Min Thigh Energy (J)	40.83 (6.25)	40.8 (6.38)	41.46 (6.02)	41.4 (5.89)	41.16 (5.79)	40.86 (5.84)	40.55 (5.86)	40.25 (5.5)

Effective vertical stiffness (Table 4.6) is the result of dividing the peak vertical ground reaction force by the downward displacement of the COM during the stance phase. The effective vertical stiffness values increased with speed for both limbs, more than doubling from the slowest to fastest speeds. The inter-limb differences were small (<1%) at all submaximal speeds, but jumped to >3% at maximal speed. These differences will be discussed further in the next section on asymmetry.

Table 4.6. Vertical Stiffness. Left (L) and right (R) Mean \pm SD stiffness values across speed.

	70		80		90		100	
	L	R	L	R	L	R	L	R
Effective Vertical Stiffness (kN/m)	28.16 (5.97)	28.88 (7.28)	35.76 (7.21)	36.4 (9.14)	47.07 (10.12)	47.42 (9.51)	67.21 (16.85)	63.72 (15.82)

4.3 Asymmetry

The difference between right and left leg performance was quantified using an adapted version of the asymmetry angle (Zifchock et al, 2006, Exell et al., 2012a), termed here as an absolute asymmetry score (AbSA). These scores assessed the absolute step to step bilateral differences of key sprinting parameters, with a score of zero indicating perfect symmetry between the right and left sides. The reader is encouraged to refer to the Methods (see Figure 3.1) regarding the non-linear relationship between AbSA scores and percent differences.

Because the sprint bouts were completed in a blocked “pyramid” order (Table 3.1), submaximal speed trials were completed both before and after bouts of maximal sprinting. To test for potential effects of fatigue in the trials after the maximal speed efforts, the interaction between speed and trial was examined. Trial and speed did not interact significantly ($p < 0.05$) for any parameter tested, indicating that participants

moved in a similar manner during bouts of the same speed regardless if the bout was completed before or after maximal speed runs.

AbSA scores for the temporal gait parameters were calculated and compared across speed conditions (Table 4.7). There were no significant main effects of speed for any of the temporal asymmetry variables. The largest asymmetry values were seen in flight time (4.15 and 3.30 at the slowest and fastest speeds respectively). Stride length asymmetries averaged 0.41, while those of step length and step time were slightly larger at 0.77 across speeds. Stance time asymmetries were 1.23 and 0.91 at the slowest and fastest speeds respectively, but again there was no significant ($p < 0.05$) trend with speed. Finally, the bilateral differences in the timing of peak eccentric knee flexion muscle moment power averaged 0.84 across speeds.

Table 4.7. Temporal Asymmetries. Average Absolute Asymmetry Scores (AbSA) for all tested temporal parameters (Mean \pm SD) across all speed conditions, and the probability of differences in asymmetry due to speed (p).

ASYMMETRY SCORE					
MEAN (\pm SD)	70	80	90	100	p
Stride Length	0.41 (0.35)	0.37 (0.39)	0.40 (0.37)	0.44 (0.65)	0.910
Step Length	0.85 (1.03)	0.69 (1.00)	0.84 (1.05)	0.68 (0.58)	0.704
Step Time	0.85 (1.03)	0.69 (1.00)	0.84 (1.05)	0.68 (0.58)	0.704
Flight Time	4.15 (5.10)	2.77 (3.79)	3.25 (3.76)	3.30 (4.38)	0.549
Stance Time	1.23 (1.88)	0.82 (1.10)	1.18 (1.72)	0.91 (0.89)	0.580
Peak Knee Flexion Power Timing	0.77 (0.52)	0.86 (1.93)	1.00 (3.20)	0.74 (0.76)	0.740

* Significant ($p < 0.05$) difference between speeds.

† Significant ($p < 0.05$) Linear relationship between speed and AbSA.

Asymmetry scores for the kinetic gait parameters are reported as speed condition means and standard deviations in Table 4.8. For the kinetic parameters, effective vertical stiffness and A/P brake impulse asymmetries changed significantly ($p < 0.05$) with speed. Effective vertical stiffness asymmetries increased from 3.87 in the 70% speed condition

to 8.11 at maximal speed in a linear fashion ($r^2=0.97$). In contrast, A/P brake impulse asymmetries did not change systematically with speed increases, with the largest disparity seen between the 90% maximal and maximal speed conditions (4.81 compared to 7.14, respectively). The largest kinetic asymmetries were seen in the horizontal and vertical propulsive ground reaction force impulses, where even at the slowest speeds the bilateral differences were 15.09 and 12.72 respectively. Both A/P propulsive and vertical impulse absolute asymmetry scores almost doubled as speed increased to maximal but were not significantly different across speed conditions. Both maximum hip power and minimum knee power asymmetries remained consistent across speeds, with scores averaging 4.57 and 4.11 across speeds.

Table 4.8. Kinetic Asymmetries. Average Absolute Asymmetry Scores (AbSA) for all tested temporal parameters (Mean \pm SD) across all speed conditions, and the probability of differences in asymmetry due to speed (p).

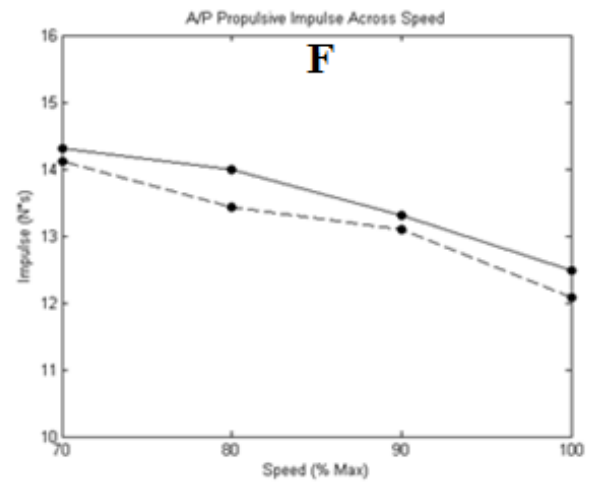
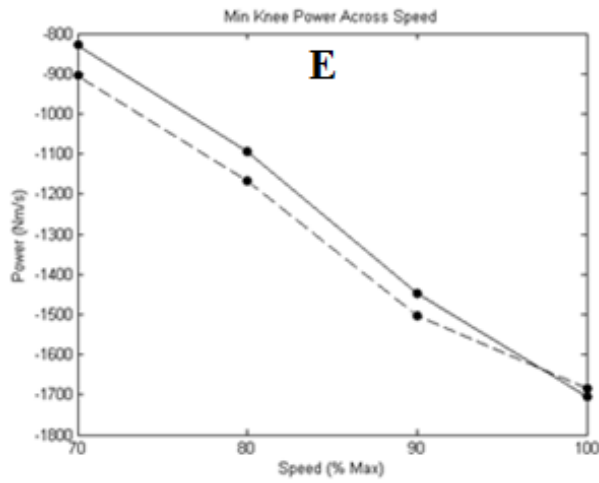
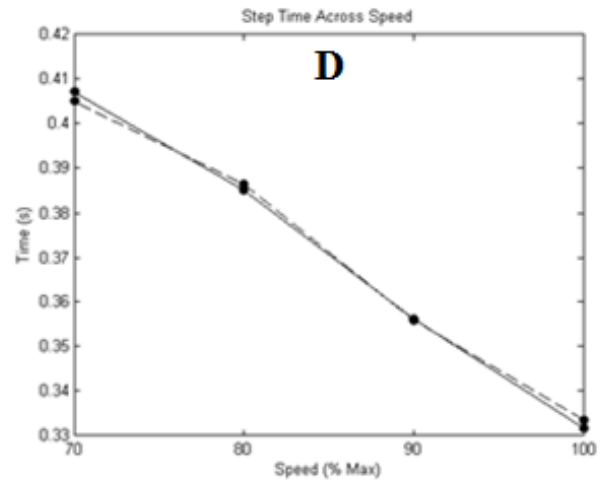
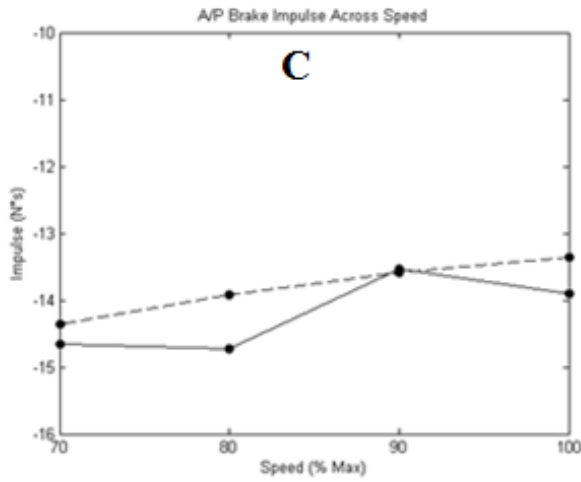
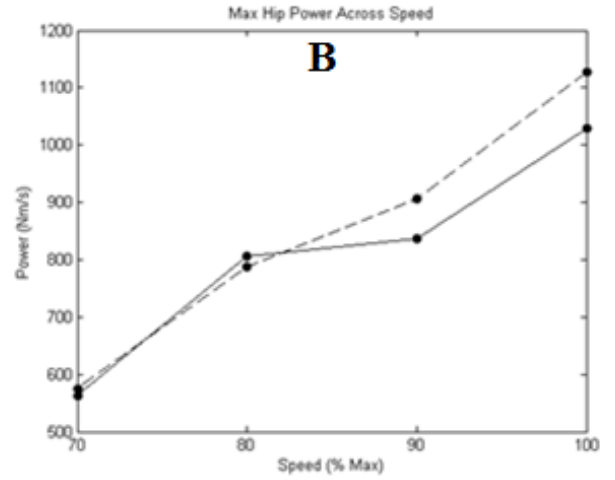
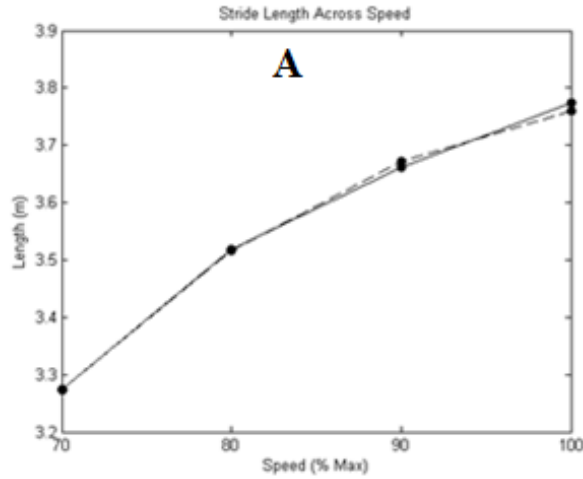
ASYMMETRY SCORE MEAN (\pm SD)	70	80	90	100	p
Peak Hip Flexion Power	4.55 (2.81)	5.41 (7.26)	4.65 (5.45)	3.68 (4.93)	0.775
Peak Knee Flexion Power	4.33 (4.33)	3.88 (3.41)	4.5 (4.77)	3.73 (4.11)	0.246
Vertical Propulsive Impulse	7.99 (15.16)	9.22 (16.51)	14.49 (20.4)	12.72 (18.88)	0.317
A/P Propulsive Impulse	8.21 (14.99)	9.98 (16.36)	13.44 (19.23)	15.09 (24.79)	0.246
A/P Brake Impulse*	5.17 (3.85)	6.06 (4.28)	4.81 (4.2)	7.14 (7.26)	0.032
Effective Vertical Stiffness*†	3.87 (2.72)	5.88 (4.34)	6.54 (4.97)	8.11 (6.06)	0.001

* Significant ($p<0.05$) difference between speeds.

† Significant ($p<0.05$) Linear relationship between speed and AbSA.

Temporal and kinetic ensemble average values from both right and left legs were examined in order to aid in the interpretation of the asymmetry scores (Figure 4.2). As expected, step and flight times all decreased with increasing speed (Figure 4.2 C and G) as did A/P braking impulses and both A/P and vertical propulsive impulses (Figure 4.2 C, F and I). Effective vertical stiffness in both legs increased with speed as did peak positive

hip and peak negative knee powers. Average A/P propulsive impulse generated by the right limb exceeded that by the left at all speeds (Figure 4.2 F). Noticeable differences between limbs were seen at all speeds for vertical propulsive impulse, with left dominating right at the three submaximal speeds, but the reverse being true at maximal speed (Figure 4.2 I). This pattern was mirrored by peak negative knee muscle power (Figure 4.2 E).



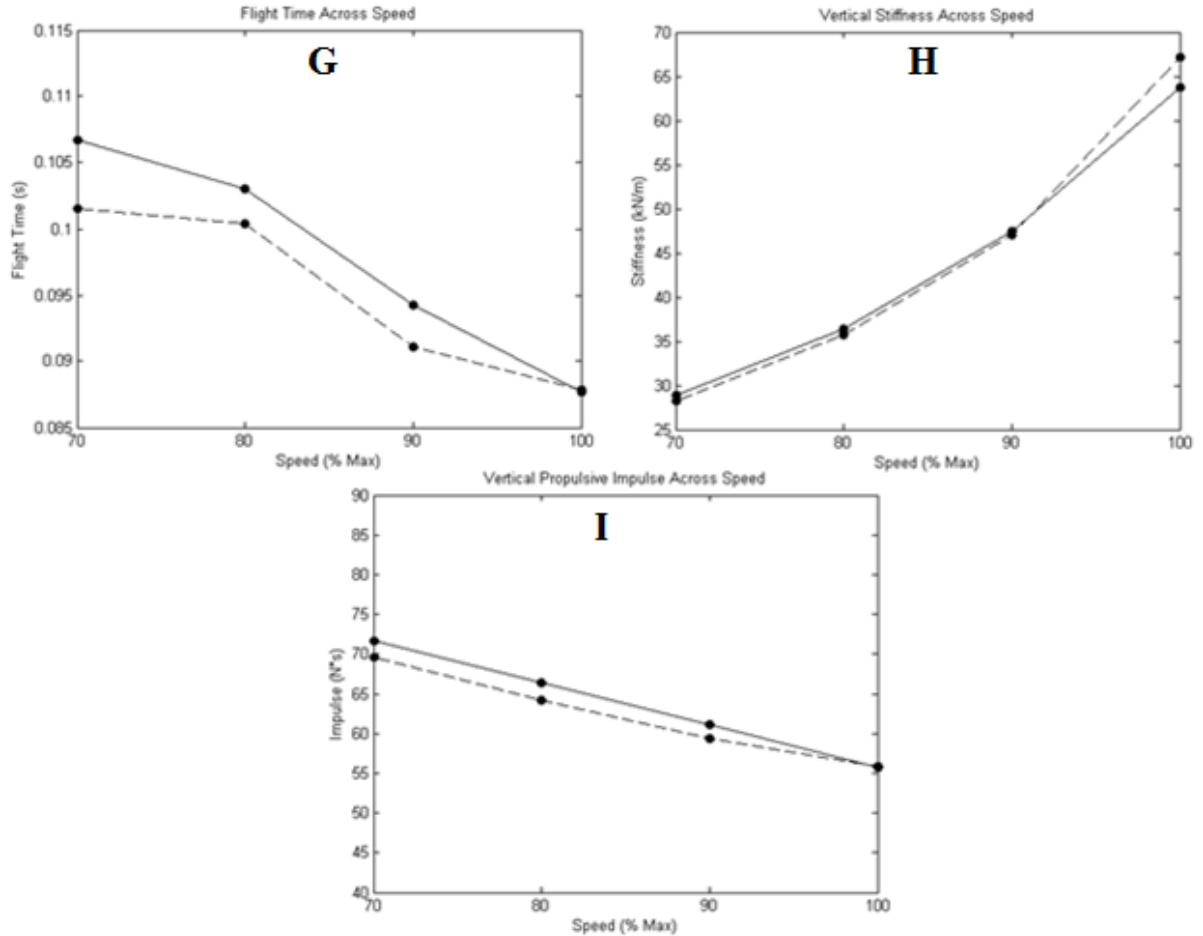


Figure 4.2. Parameter Averages Across Speed. Ensemble average left (dashed) and right (solid) Stride lengths (A), Max Hip Muscle Moment Powers (B), A/P Braking Impulses (C), Step Times (D), Min Knee Muscle Moment Powers (E), A/P Propulsive Impulses (F), Flight Times (G), Effective Vertical Stiffness (H) and Vertical Propulsive Impulses (I) across speed (70%-100% Maximal).

4.4 Effective Vertical Stiffness Characteristics

Upon further examination, the pattern of change for the absolute asymmetry scores did not agree with the magnitudes of the differences between the average values (AvD) at each speed for effective vertical stiffness (Table 4.9). As submaximal speed increases from 70 to 90% of maximal, the absolute step-to-step differences (AbSA) increased from 3.87 to 6.54, while the average inter-limb differences (AvD) decreased from 0.71 to 0.35. However, both measures reached their highest values during the maximal speed condition.

Table 4.9. Absolute and Average Stiffness. Effective Vertical Stiffness absolute asymmetry scores (AbSA), average inter-limb difference (AvD) and percent inter-limb difference based on AbSA and AvD.

Effective Vertical Stiffness Characteristics	70	80	90	100
AbSA	3.87	5.88	6.54	8.11
% Difference AbSA	11	17	19	23
AvD (kN/m)	0.71	0.64	0.35	3.49
% Difference AvD	2.5	1.7	0.7	5.2

Absolute differences will always be larger than average ones as there is no directional component to the AbSA calculation. Small average differences result from situations in which the left limb values were larger than the right about the same number of times as the converse situation (e.g. 70%, 80% and 90% speed conditions). As speed increased from 70% to 90% maximum, the step-to-step differences in effective vertical stiffness grew, but both limbs managed to oscillate around similar average values. When average differences are large (100% speed condition), the left limb's values more consistently exceeded those of the right limb (Figure 4.2 H), creating the discrepancy in average effective vertical stiffness.

A closer examination of some determinants of effective vertical stiffness at maximal speed also revealed inter-limb differences (Table 4.10). On average the maximum vertical forces experienced during left and right foot contacts were equivalent, but the vertical COM displacement during right foot contacts was greater than during left ones. Joint angular displacements during the propulsion phase were similar (<6% different between limbs).

Table 4.10. Stiffness Characteristics. Average Left (L) and right (R) limb Effective Vertical Stiffness characteristics during maximal speed sprinting.

Maximal Speed Characteristics	Left	Right
Effective Vertical Stiffness (kN/m)	67.21	63.72
Max Vertical Force (N)	1907	1907
COM Displacement (m)	0.031	0.033
Propulsive Phase Ankle Angular Displacement (deg)	44.49	44.4
Propulsive Phase Knee Angular Displacement (deg)	23.92	22.57
Propulsive Phase Hip Angular Displacement (deg)	41.86	41.01

Although subjects exhibited nearly identical angular displacements between limbs, bilateral differences existed with respect to the relative angular positions of the three lower limb joints during the propulsion phase of stance. Subjects maintained more dorsi-flexed ankle and extended hip positions during left limb contacts than right ones throughout the propulsive phase (Figure 4.3 A and C). Relatively similar knee angular positioning was observed for the first half of propulsion, though the left limb extended more than the right during terminal stance into toe off (Figure 4.3 B).

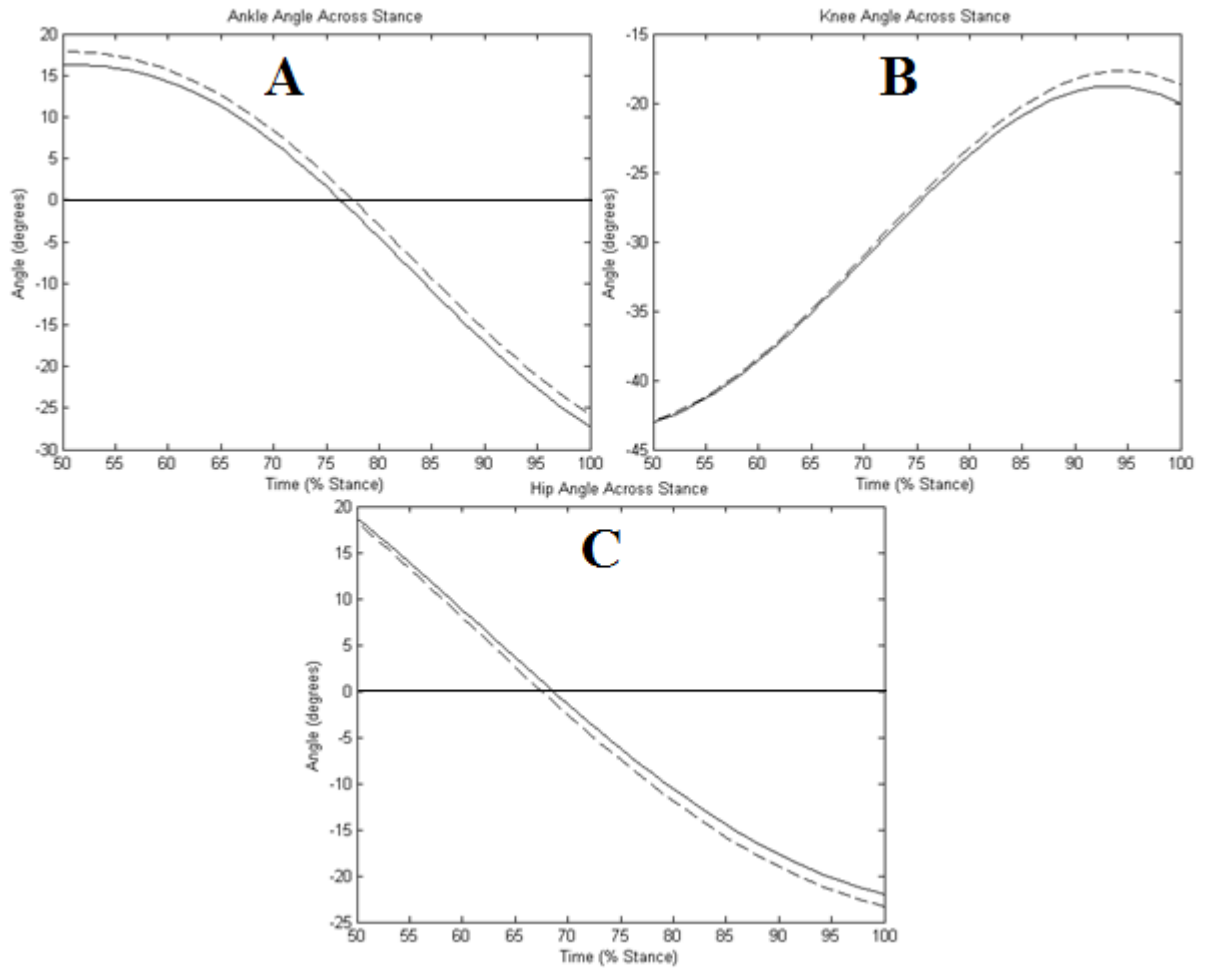


Figure 4.3. Joint Angles During Propulsion. Ensemble average left (dashed) and right (solid) ankle (A), knee (B) and hip (C) angles during stance. 100% of stance indicates toe-off, while 50% approximates the beginning of the propulsion phase. Positive values indicate a relatively dorsi-flexed ankle, an extended knee and flexed hip compared to anatomical position. Negative values indicate a relatively plantar-flexed ankle, a flexed knee and extended hip relative to anatomical position.

CHAPTER 5

DISCUSSION

Previous research has identified certain kinetic limitations to maximal sprint speed (Weyand et al., 2010; Chapman and Caldwell, 1983a; Miller, Umberger and Caldwell, 2012). However, these studies were unilateral in nature and mostly limited to either stance (Weyand et al., 2001) or swing (Chapman and Caldwell, 1983b) phase alone. Considerable evidence exists for natural kinetic asymmetries during both submaximal (Zifchock et al., 2008) and maximal speed (Exell et al., 2012b) running, but it is unclear how the magnitudes of these asymmetries change from submaximal to maximal speed. Therefore, the purpose of this investigation was to examine kinetic asymmetries of key parameters associated with limitations to maximal sprint speed, with specific interest in how these asymmetries change with increasing speed. Of all parameters examined, only effective vertical stiffness asymmetries significantly increased with speed. Therefore, the hypothesis that asymmetries in all key sprinting parameters would be significantly different at maximal speed than at all submaximal speeds is rejected.

5.1 Comparison with Previous Research

The subjects in this study were competitive female athletes, experienced with maximal-speed sprinting and treadmill running. Previously, scientists have examined sprint-related phenomenon using average- and highly-trained males and females with top speeds ranging from 6.77 m/s (Belli et al., 1995) to 9.20 m/s (Weyand et al., 2010), and our average top speed of 7.72 m/s is within this range. The changes in stride lengths and frequencies of ~ 13% and ~ 20% respectively from the slowest to fastest speeds were

similar to those found by Brughelli and colleagues (2011) in their examination of semi-professional Australian Rules Rugby athletes. Furthermore, the stride frequencies of our subjects (average 2.056) were almost identical to those of other trained females sprinting at similar velocities (Chapman and Caldwell, 1983b). The trends with running speed depicted in Figure 4.1 are similar to those in the literature, although the increases (~0.5 m) in stride length from 70% to maximal speed were not apparent in data from Mero and Komi (1978). However, our speed range is smaller, thereby emphasizing differences in these parameters from 70% to maximal speed. Presented on the same axes scales as previous work, the differences from 70% to maximal speed would not be as apparent. In contrast, our reported increases in stride frequency were not as apparent as in previous literature. Perhaps the adaptation of these athletes to treadmill running was to increase stride length at the higher speeds as opposed to stride frequency. Most of our subjects played team sports, so they may differ from trained sprinters by depending more on changes in stride length rather than stride frequency to attain top speed.

Peak vertical forces linearly increased with speed as in Brughelli et al (2011) and Weyand and colleagues (2010). During maximal speed running, the peak vertical forces were smaller in magnitude for the current subjects ($1907 \pm 224.86\text{N}$ at top speed) than Brughelli and colleagues' (2011) subjects ($1983.7 \pm 271.9\text{N}$), likely due to the smaller average masses and slower average speeds of our subjects. Finally, net hip and knee muscle moment powers agree with those measured during late swing phase by Chapman and Caldwell (1983b) for both polarity and magnitude (hip $>2000\text{W}$ and knee $\sim -1000\text{W}$) at similar speeds ($\sim 7.6\text{ m/s}$).

There are several theories in the literature regarding the nature of how runners attain their maximal speed. Relative to leg length, experienced sprinters take longer strides than those of non-sprinters, therefore some authors postulate that maximal speed is dependent on the impulse generated vertically to propel the runner into longer strides (Weyand et al., 2001). A different theory is that speed is limited by the need to manage the mechanical energy of the swinging limb for rapid recovery and proper positioning of the foot for the subsequent stance phase. Advocates for this theory maintain that muscle moment powers at the hip and knee are limited in their ability to generate and absorb energy during initial and terminal swing phase (Chapman and Caldwell, 1983b). The present data corroborate aspects of both views. First, peak vertical ground reaction forces increased with speed (Table 4.3), while stance times decreased (Table 4.2), leading to lower vertical propulsive impulses (Figure 4.2 I) and decreased flight times (Figure 4.2 G) as maximal speed was reached. Therefore, a limitation to top speed could be an inability to stay in the air longer due to the ever-decreasing impulses that oppose gravity. There is no plateau of either peak hip or knee muscle moment powers as running speed progresses to maximal in this study (Figure 4.2 B and E). However, the previous work in this area was completed on a much tighter range of speeds, closer to maximal (Chapman and Caldwell, 1983b; Swanson and Caldwell, 2005) than the speeds in this study. In fact, Swanson and Caldwell (2005) noted that the hip and knee muscle powers exhibited by their subjects at top speed were different from their ninety percent but similar to their ninety five percent conditions. Therefore, the patterns observed by those authors may only occur when examining smaller incremental differences as maximal speed is approached. However, our results support the idea that an inability to slow the foot in

late swing may limit maximal speed. Foot energy and thus velocity was higher at foot contact with increasing speed (Figure 4.7 B), which could slow the forward velocity of the athlete's CoM at touchdown (Mann, 1981).

While support exists for both maximal velocity limitation theories, there is also considerable evidence suggesting that inter-limb interactions affect performance (Vagenas and Hoshizaki, 1992; Exell et al., 2012a; Bartlett et al., 2007). Exact replication of a cyclical motion is impossible due to the interactions with ever-changing environmental conditions (Bartlett et al., 2007; Vagenas and Hoshizaki, 1992), and multiple internal degrees of freedom that must be coordinated (Bernstein, 1967; Bartlett et al., 2007). Inter-limb variability has been found to increase with speed (Belli et al., 1995; Swanson and Caldwell, 2005), and kinetic asymmetries tend to be larger than kinematic and temporal ones at maximal speed (Exell et al., 2012a). The results of the current study agree with those of Belli and colleagues (1995) as the highest standard deviations exhibited by the stance phase kinetic parameters were observed at maximal speed (Table 4.6). Furthermore, the kinetic asymmetry scores exceeded those of the temporal parameters at the three fastest speeds (compare Table 4.6 to Table 4.5), in agreement with Exell et al. (2012a). In addition, there were differences of ~11% between limbs observed across all speeds for swing phase kinetic parameters (Table 4.6), in agreement with findings of Swanson and Caldwell (2005). Failure to effectively coordinate the lower limbs during swing (as indicated by asymmetry and increased variability) could hinder effective force application during stance and eventually limit maximal sprint speed (Swanson and Caldwell, 2005).

Various research groups have proposed interpretations of asymmetry scores. Some groups consider inter-limb differences significant only if the asymmetry score is greater than 10% (Zifchock, Davis and Hamill, 2006; Zifchock et al., 2008), while others suggest that asymmetry must exceed intra-limb variability to be considered significant (Exell et al, 2012a; Exell et al., 2012b). Analyzing the absolute asymmetry scores in concert with other kinetic and kinematic averages can describe the asymmetry landscape better than simply viewing either alone. Waveform or trend symmetry has been advocated for describing general changes in motion, but perhaps other techniques such as principal components analysis and various correlation algorithms could be implemented to assess significance for discrete kinetic data (Crenshaw and Richards, 2006). It is the opinion of this author that it is not enough to simply state that limbs “look different”, and more work needs to be done to determine at which level asymmetries really describe true differences between limbs. Perhaps the nature as opposed to simply the amount of variability needs to be taken into account. From the current data, we can say that inter-limb differences in effective vertical stiffness grew significantly ($p < 0.05$) as subjects increased speed to maximal.

5.2 Asymmetry Data

The asymmetry data in Tables 4.5 and 4.6 and the individual panels of Figure 4.2 demonstrate that patterns of change across speed are consistent between limbs for some but not all parameters. We reported asymmetries as absolute AbSA scores, although these scores are somewhat difficult to interpret because of their non-linear nature (Figure 3.2). Appendix G contains absolute AbSA asymmetry scores and their corresponding between-limb percentage differences. The absolute asymmetry index used here is robust because it

results in the same score regardless of which leg's value is larger, which removes the need to designate a "normal" limb. In the discussion that follows, inter-limb differences will be described using percentage differences where appropriate (See Appendix G).

Four scenarios can be identified to encompass and categorize the range of variable and asymmetry responses displayed by our subjects. In the first scenario, a parameter (e.g. Peak Negative Knee Power Timing) exhibits relatively unchanging average values and consistently low asymmetry scores across all speeds (Figure 5.1). Peak eccentric knee muscle moment power occurs at about eighty percent of the gait cycle regardless of speed in both limbs. In addition, the difference in timing between limbs (i.e. the asymmetry in this timing) does not significantly change with speed.

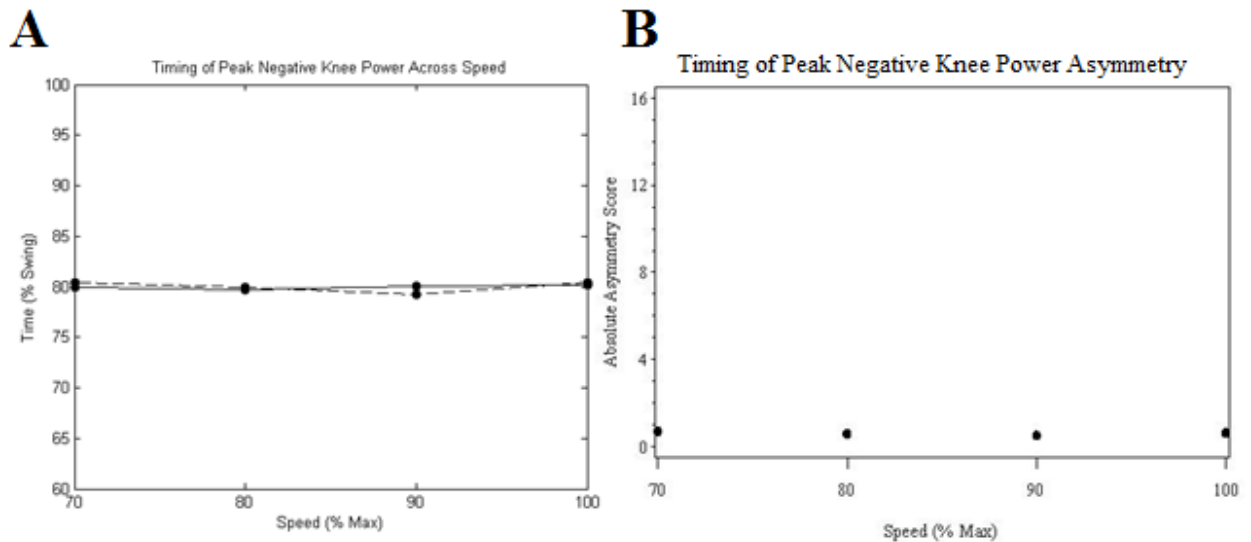


Figure 5.1. Asymmetry Scenario 1. Timing of Peak Negative Knee Power during terminal swing left (dashed) and right (solid) (A) and asymmetry (B) changes with speed.

The second scenario represents parameters (e.g. stride length) for which the average values change, but the level of asymmetry is consistently small across all speeds (Figure 5.2). This scenario encompasses stride and step lengths and stride, step, stance and swing times.

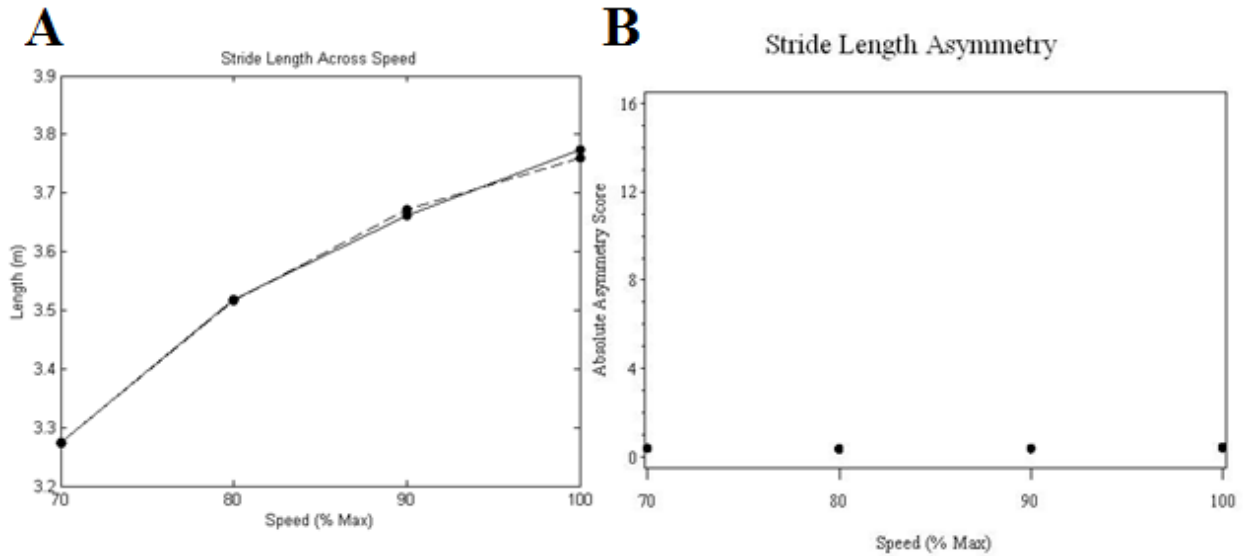


Figure 5.2. Asymmetry Scenario 2. Stance time left (dashed) and right (solid) (A) and asymmetry (B) changes with speed.

The third scenario represents parameters for which there were consistent qualitative differences between right and left limb values, even though the magnitudes of the values changed as speed increased (Figure 5.3). For example, it appears as though subjects produced higher A/P propulsive impulse with their right limb than their left at all speeds (not statistically tested), but the impulse generated by both limbs decreased as they sprinted faster. Even though the magnitudes of those between-limb asymmetries increased by ~10% from the slowest to fastest speeds, these increases were not significantly different between speeds. In addition, vertical propulsive impulse, minimum knee muscle moment power, maximal hip muscle moment power and flight times fit into this category based on qualitative inspection of the average values and bilateral asymmetries exceeding ten percent across speeds.

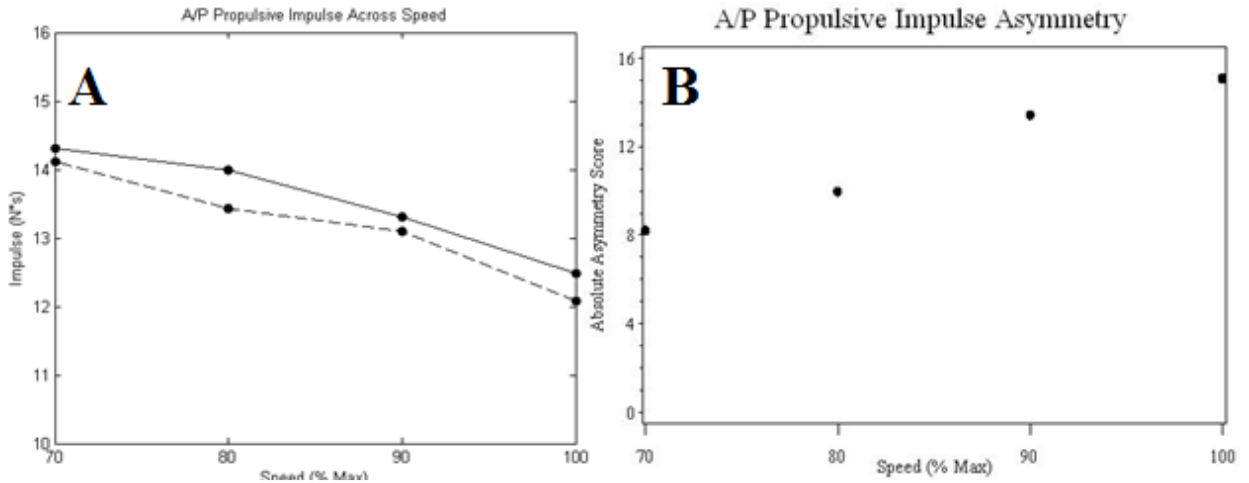


Figure 5.3. Asymmetry Scenario 3. Average A/P propulsive impulse for left (dashed) and right (solid) limbs (A) and asymmetry (B) changes with speed.

The final scenario involves parameters (e.g. effective vertical stiffness) for which both the average values and the level of asymmetry change across speeds (Figure 5.4). The subjects displayed increased vertical stiffness during foot contacts on both limbs as speed increased. In addition, the difference in vertical stiffness between limb contacts (i.e. asymmetry) increases with speed. Effective vertical stiffness and A/P brake impulse were the only tested parameters for which this scenario applied.

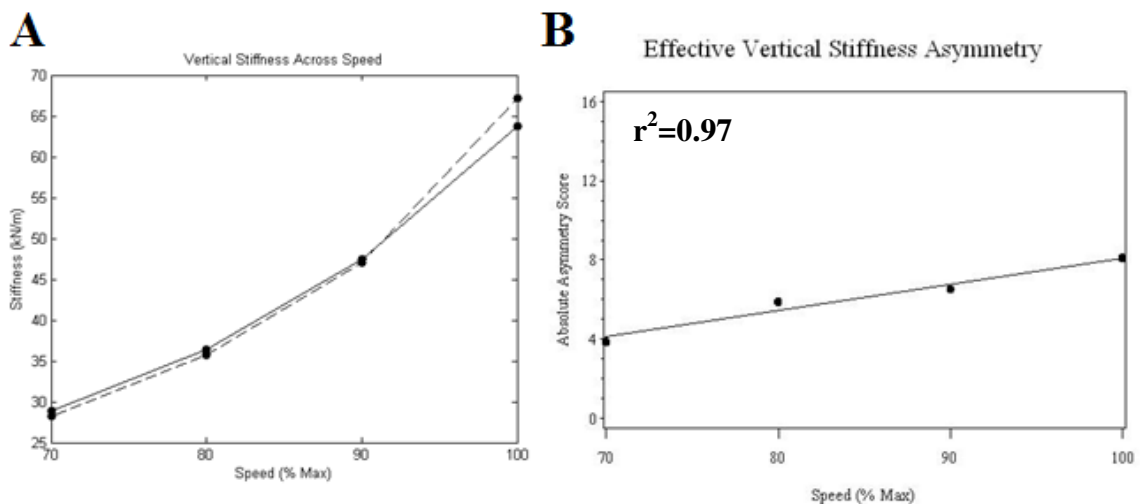


Figure 5.4. Asymmetry Scenario 4. Effective Vertical Stiffness left (dashed) and right (solid) (A) and asymmetry (B) changes with speed.

5.3 Effective Vertical Stiffness

By modelling the lower limb as a linear mass-spring system, mechanical spring constants can be used to describe the resistance of vertical COM motion to a corresponding vertical force, known as effective vertical stiffness (McMahon and Cheng, 1990). High stiffness values indicate that the COM undergoes little vertical displacement when confronted with an applied vertical force. It has been reported that higher effective vertical stiffness values are associated with increases in speed (McMahon and Cheng, 1990), stride frequency (Farley and Gonzalez, 1996) and sprinting ability (Mero and Komi, 1986). Our data agree with these studies, as effective vertical stiffness increased from seventy percent to maximal speed in a manner qualitatively similar to the results of McMahon and Cheng (1990) (Figure 5.5).

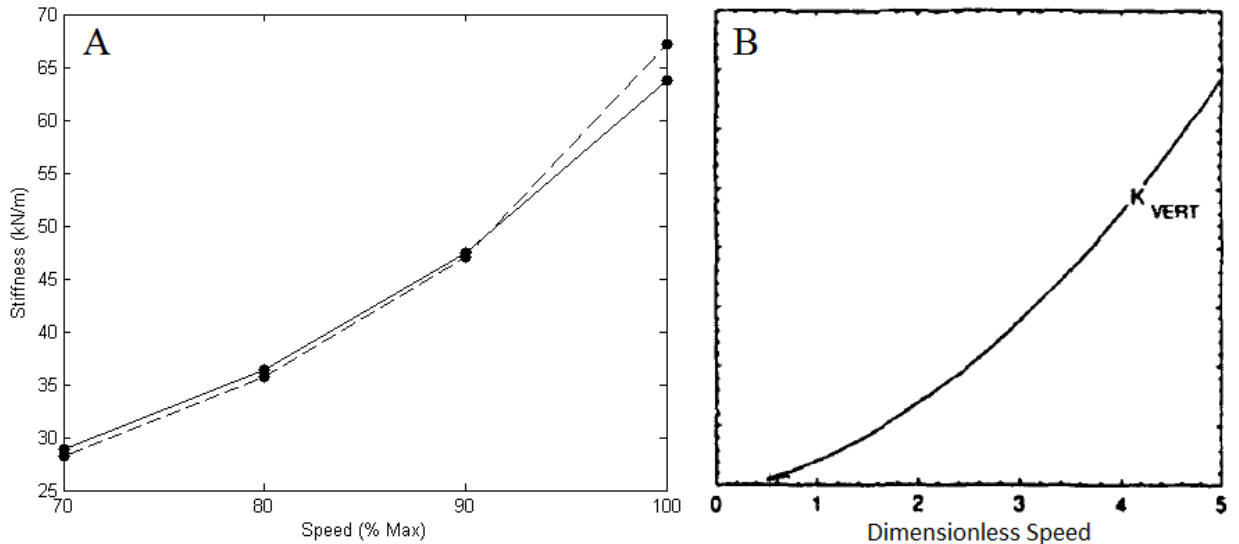


Figure 5.5. Stiffness Comparison. Effective vertical stiffness values from the current study (A) and from McMahon and Cheng (1990) (B).

McMahon and Cheng (1990) presented stiffness values in relation to dimensionless speed (Figure 5.5 B) because they tested both bipedal and quadrupedal animals, with speed 5 and 1 representing the fastest and slowest speeds respectively. For

our sprinters, the bilateral stiffness values are close in magnitude at the submaximal speeds but diverge at maximal speed, with effective vertical stiffness larger for left (67.21 kN/m) versus right limb (63.72 kN/m) contact phases. At submaximal speeds, the step-to-step inter-limb differences grew with increases in speed but the opposite effect was witnessed for average inter-limb differences (Table 4.9). The step-to-step scores are non-directional while the average scores have a directional component; therefore the interpretation is that left and right values grew increasingly different from each other as speed increased, although they oscillated around similar average values. At maximal speed, the effective vertical stiffness values during left limb contacts were ~23% larger than during right contacts for absolute asymmetry scores, and ~5% larger for average asymmetry scores (Table 4.9). This indicates that the left limb tends to be stiffer than the right during stance phase at top speed more often than the converse situation.

Arampatzis and colleagues (2001) found that jump height (and thus power output) in vertical jumping was maximized not at peak stiffness, but at some submaximal stiffness level, suggesting that the ability to quickly generate mechanical energy declines as limbs stiffen beyond some “optimal level”. In our sprinting data, there is no indication that higher stiffness compromised left limb mechanics as the vertical propulsive impulse was on average higher during left foot contacts than right ones (55.87 N*s versus 55.71 N*s respectively) (Figure 4.2 I). In fact, by increasing the effective vertical stiffness during left foot contacts, the subjects appeared able to curb the decrease in vertical propulsive impulse as speed progressed to maximal. Weyand et al. (2001) determined that applying greater forces in opposition to gravity lead to greater flight distances and thus faster sprinting performance. Furthermore, applying these necessary forces in less

time could result in even better sprinting performance. However, increasing vertical stiffness may compromise horizontal force production, as horizontal propulsive impulse was lower during left limb contacts compared to right (Figure 4.2 F). Others have shown that trade-offs occur at maximal speed between stride length and stride frequency (Donati, 1995), and between stride time and stride distance (Bosco, 1986). Interestingly, the right ankle is more plantar-flexed than the left at toe off (Figure 4.3 A), which may be a more advantageous position to direct force posteriorly rather than vertically.

5.4 Stiffness Asymmetries and Sprint Speed Limitations

The vertical motion of the COM during human running has been compared to that of a bouncing ball (Cavagna, 1976; Heglund et al., 1982) and can be described by a linear spring-mass model (McMahon and Cheng, 1990). When the CoM moves in a downward direction during initial stance, energy is stored in the elastic tissues of the extensor muscles. This energy is released during the propulsive phase, supplementing the active muscular work performed against gravity. This passive storage and release cycle enables animals to run both faster and more efficiently than if all propulsion was performed by active musculature (Cavagna et al., 1976).

With higher stride frequencies at faster speeds the stance phase is shortened, with less time to produce large forces in opposition to gravity (Weyand, 2001). To offset the reduced time on the ground, the effective vertical stiffness of the support limb is increased, resulting in quicker “bounces” (Farley et al., 1993; Farley and Gonzalez, 1996). Not surprisingly, effective vertical stiffness has been implicated as a primary limiter of maximal running speed (Farley, 1997).

Effective vertical stiffness asymmetries may contribute to limiting maximal sprint speed by creating vertically unstable gait at high speeds, and inhibiting athletes from further increasing stride frequency. Stable gait refers to the tendency of a system to return to its prescribed path or condition after a perturbation (Wagner and Blickhan, 1999). The prescribed path to which we are referring is the cyclical vertical oscillation of the COM, and the perturbations are the ground reaction forces experienced by the runner during stance phase. In addition, previous research indicates that symmetrical stance phases with respect to the vertical axis would result in cyclical (stable) trajectories of the COM, and that deviations from symmetry likely will not (Seyfarth et al., 2001). Consequently, asymmetrical limb stiffness could upset cyclical COM motion, creating an unstable running situation that would likely degrade performance.

Increased joint stiffness was necessary for two- and three-segment robotic models to maintain running stability at faster speeds (Rummel et al., 2008). This suggests that increasing stiffness bilaterally could allow symmetrical and stable gait to continue at higher speeds. For our subjects the step-to-step stiffness differences were substantial (>10% between limbs) and increased with speed, but similar average bilateral stiffness values (<3% difference) were observed during conditions up to 90% max speed (Table 4.6). However, subjects were unable to bilaterally match their step-to-step (~23% different) or between-limb average (~5% difference) effective vertical stiffness at maximal speed, indicating less stability in their gait. Perhaps there is some threshold asymmetry associated with gait instability, thereby limiting the maximal attainable speed. Further investigation is needed to determine the nature of this potential threshold.

Previous research has indicated that runners attempt to overcome unstable stance phase conditions by adjusting horizontal foot velocities and leg angles during swing phase (Seyfarth et al., 2003), further limiting the ability to increase COM velocity. An examination of the relationship between stride length and frequency and the actual effective vertical stiffness values during maximal speed running (Tables 4.2 and 4.10) suggests that this situation may have occurred. Average effective vertical stiffness values were larger (Table 4.10) and stride frequencies were higher during left limb strides than right ones at top speed (Table 4.2). Subjects increased the length of their strides during right limb contacts (Table 4.2), possibly adjusting their foot velocities and positions in the process. This may have been a back-up strategy used only after exhausting their ability to bilaterally increase stride frequency equally (as they did up to 90% max speed).

The mechanical demands of increasing stride frequency are much different than those of increasing stride length (Donati, 1995; Hunter, Marshall and McNair, 2004) because increasing stride frequency necessitates a more rapid repositioning of the limbs, while increasing stride length requires larger ground reaction force impulses to propel a runner into *longer* flight phases. Thus, increasing stride frequency results in less flight time, while greater stride lengths tend to produce longer flight times. Clearly stride frequency and stride length-mediated increases in sprint speed require different mechanical strategies, and the kinematic asymmetries observed in Figure 4.3 indicate that our subjects may have been attempting to coordinate both at maximal speed. Previous research has shown that a more forward-oriented lower body position results in better horizontal acceleration (Kugler and Janshen, 2010), and perhaps it is this mechanism by which our athletes attempted to counterbalance the discrepancy in stiffness. Although the

angular displacements were similar (<6% difference) between limbs, our subjects maintained more plantar-flexed ankle, flexed knee and flexed hip positions during right foot contacts than left ones during propulsion (Figure 4.3 A, B and C). This combination of joint angles would tend to orient the lower limb in a more anterior position. This kinematic adjustment could have been made in an effort to direct force more posteriorly during propulsion on their right limbs, thus increasing stride length during right limb strides. If subjects attempted to use both strategies simultaneously, their ability to coordinate their movement may have been further compromised. Thus, vertical stiffness asymmetries could render gait unstable, limit the ability to further increase stride frequency, and therefore limit an athlete's maximal running velocity.

The effective vertical stiffness measure reflects the mechanics of the entire lower extremity on CoM motion, but what contribution is made by individual joints? Joint stiffness is calculated by dividing peak net joint moment by angular displacement. Examination of the peak net extensor stiffness of the three lower limb joints during the 100% speed condition does not explain the inter-limb differences in effective vertical stiffness (Figure 5.6). Although not tested statistically, the subjects exhibit equivalent or higher joint stiffness values during right limb propulsion than left, and therefore cannot explain the opposite trend observed for effective vertical stiffness. Consideration of joint stiffness during the propulsion phase alone is not sufficient to capture the whole-body trend for vertical stiffness, at least in part because angular joint motion produces both vertical *and* horizontal linear motion. Another possibility is that motion in other planes (e.g. frontal plane pelvic tilt) could displace the COM vertically. Consequently, more

research in different planes and at the joint level needs to be completed to deduce the mechanism behind the asymmetrical stiffness exhibited by our athletes.

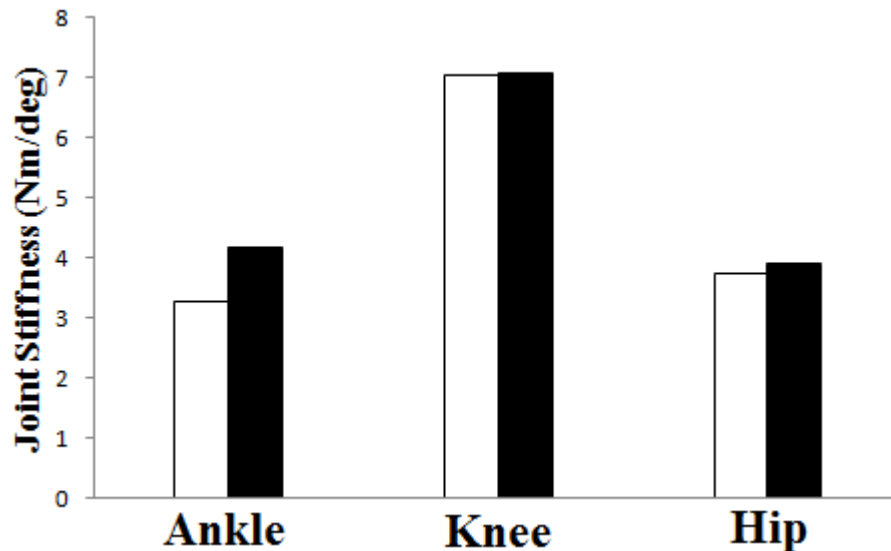


Figure 5.6. Joint Stiffness. Ensemble average Peak Joint Stiffness during the propulsive portion of stance phase during the 100% condition. White and Black bars indicate left and right values, respectively.

5.6 Practical Implications

These results have many implications in the realms of sport science and rehabilitation. First, effective vertical stiffness asymmetries at top speed could result in mechanical and coordinative compensations by athletes as they attempt to maintain forward CoM motion. Such motion compensations will in general slow the maximal forward speed attainable by the runner.

Effective vertical stiffness imbalances of 6.5% have been reported in healthy male Australian Rules Football athletes running at 80% of their maximal aerobic velocity on a treadmill (Brughelli et al., 2010). If strength and conditioning coaches and trainers were able to train athletes to maintain stiffness symmetry longer, it is plausible that their maximal speed would improve as well. Additionally, pure symmetry may not be a practical training goal for athletic therapists and trainers because some degree of

asymmetry was observed in every temporal, kinematic and kinetic variable examined. This brings up the issue of how much asymmetry is too much? One position is that asymmetries should be monitored and attempts made to decrease them when there is a potential performance improvement. If the nature of the asymmetry is strength-related, specific strength training can be targeted to reduce bilateral strength deficits and thereby improve limb motion symmetry. From a clinical perspective, asymmetries may not lead directly to (nor indicate) the presence of an injury. Given certain anatomical asymmetries, athletes may be forced to move differently from step to step. Even if no anatomical asymmetry is present, differentiating between asymmetry and movement variability in each leg is important. Clinicians could simply be witnessing movement variability, and caution should be used when attempting to alter someone's movement patterns. Furthermore, "treating" asymmetry may actually be ill-advised since the motor patterns used naturally by athletes are executed for a reason, and may not represent a compensatory response.

5.7 Study Limitations

As with all experimental studies, the study design has limitations that should be considered. A force-instrumented treadmill was used to examine bi-lateral asymmetries over multiple consecutive steps. Treadmill running differs from over ground running because on a treadmill the ground moves under the runner rather than remaining stationary. However, a recent study concluded that sagittal plane kinematic patterns and discrete values did not differ significantly between the two running conditions (Fellin, Manal and Davis, 2012). Our runners completed trials in a fixed order (a "speed pyramid") because previous research indicated that it was neither beneficial nor practical

to force subjects to complete a maximal speed trial as their first trial (Hamill et al., 1983; Thelen et al., 2005). Trial order did not interact significantly ($p < 0.05$) with speed for any parameter tested. Kinematic events were used to identify foot strikes and toe offs as opposed to force plate threshold identification because of greater noise in the force treadmill signals compared to fixed-in-ground force plates. The algorithms for computing these events have been validated for both walking (Hreljac and Marshall, 2000) and running (Hreljac and Stergiou, 2000). Although sprinters do not typically strike the ground heel-first, these kinematic algorithms are more reliable than noisy GRF signals.

5.8 Conclusion

In conclusion, bilateral asymmetries exist for a number of kinematic and kinetic parameters at both submaximal and maximal speeds. This investigation examined the relationship between speed and bilateral asymmetry of key sprint-related kinetic parameters. Effective vertical stiffness asymmetries grew as the subjects ran faster, and appeared to have an effect on the asymmetries for both vertical and A/P propulsive impulse at maximal speed. Furthermore, asymmetries in effective vertical stiffness may force runners to resort to a less stable and less coordinated gait, limiting their ability to further increase stride frequency, and thus limiting maximal speed.

APPENDIX A

MECHANICAL ENERGY CONCEPTS

Mechanical energy (E_M), the sum of the kinetic energy (E_K) and gravitational potential energy (E_P) in a system (Equation A1), is related to the motion and position of a system.

$$E_M = E_K + E_P \quad (A1)$$

$$E_k = \frac{1}{2}m \cdot v^2 \quad (A2)$$

$$E_P = m \cdot g \cdot h \quad (A3)$$

Kinetic energy (E_K) is the component of mechanical energy relating to the motion of the system, equalling one half the product of the system's mass (m) and velocity (v) squared (Equation A2). Potential energy (E_P) is the component of mechanical energy relating to the position of the system, equalling the product of mass (m), acceleration due to gravity (g) and height (h) above a reference position (Equation A3).

Measuring changes in the mechanical energy of a system is not a direct process. As a barometer measures the change in barometric pressure, mechanical work is a "pseudo-barometer" for changes in mechanical energy of a system (Equation A4). Another pertinent definition of mechanical work is the amount of energy generated, absorbed or transferred by forces (as to distinguish from energy in the form of heat).

$$W = \Delta E_K + \Delta E_P \quad (A4)$$

There are three scenarios for mechanical work in a system; positive work (which raises the energy level of the system), negative work (which lowers the energy level), and no work (no change in energy). The relationships between force, velocity, acceleration and energy are presented in Equations A5 through A8 below.

$$F = m \cdot a = m \cdot \left(\frac{\Delta v}{t} \right) = \frac{m \cdot \Delta v}{t} \quad (A5)$$

$$\text{Rearrange... } F \cdot t = m \cdot \Delta v \quad (A6)$$

$$\text{Since... } \Delta v = v_2 - v_1 \quad (A7)$$

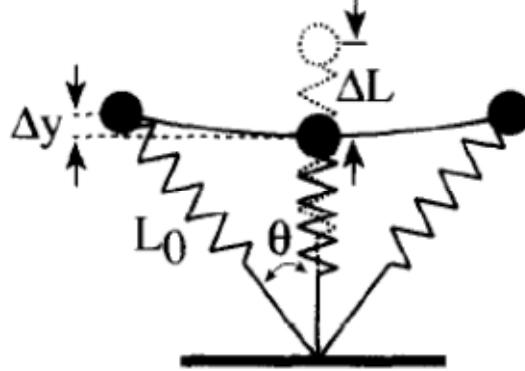
$$\Delta E_k = \left(\frac{1}{2}m \cdot v_2^2 \right) - \left(\frac{1}{2}m \cdot v_1^2 \right) \quad (A8)$$

An applied force will change the velocity (v) of an object with mass (m) proportional to the product of the magnitude of the force applied (F) and the length of time of force application (t) (Equations A5, A6 and A7). Thus, ground reaction forces change the kinetic energy (and velocity) of runners.

APPENDIX B

THE SPRING-MASS MODEL FOR RUNNING

Measurements and calculations of the input parameters to the spring-mass model adapted from McMahon and Cheng (1990) shown below.



The peak displacement of the leg spring (ΔL) was calculated from the peak displacement of the center of mass (Δy), the length of the leg spring at the instant that it hit the ground (L_0) and half of the angle swept by the leg spring while it was in contact with the ground (θ) as shown in equation (B1). Δy was measured kinematically as the vertical displacement of the sacral tracking marker from foot strike to its lowest point during stance. L_0 was measured as the distance between the greater trochanters of the hip and the 5th metatarsal head marker. The model is based on the properties of an inverted pendulum, so θ was calculated as shown in equation (B2), assuming that forward speed (u) is almost constant during the time of foot contact (t_c).

$$\Delta L = \Delta y + L_0(1 - \cos\theta). \quad (\text{B1})$$

$$\theta = \sin^{-1}(ut_c/2L_0). \quad (\text{B2})$$

Equations reproduced from McMahon and Cheng, 1990.

APPENDIX C

MECHANICAL POWER INFORMATION

SHANK

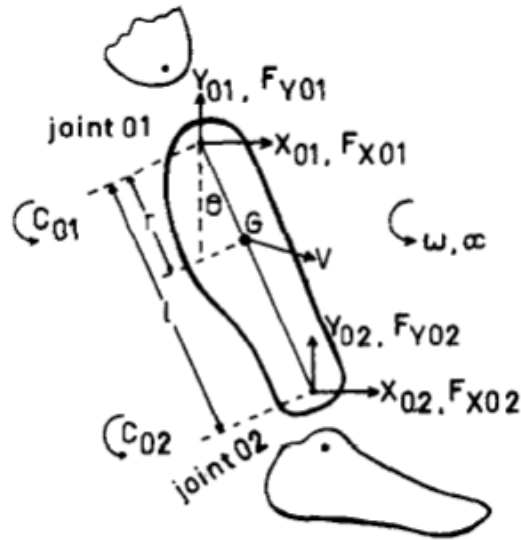


Fig. 1. Representation of the variables and subscript used in equations

α	Segment absolute angular acceleration about a transverse axis through its centre of mass in $\text{rad} \cdot \text{s}^{-2}$
C	Moment about a transverse axis through the joint in N m
F	Joint force in N
g	Gravitational acceleration, $9.81 \text{ m} \cdot \text{s}^{-2}$
h	Vertical height of segment centre of mass in m
I_G	Moment of inertia of segment about a transverse axis through its centre of mass in $\text{kg} \cdot \text{m}^2$
l	Segment length in m
m	Segment mass in kg
PF	Joint force power in W
PM	Muscle moment power in W
r	Distance between proximal joint and segment centre of mass in m
TLE	Total limb energy in J
θ	Segment angle with respect to a vertical reference through the proximal joint in radians, positive in

	front and negative behind the vertical in the direction of the run
V	Absolute translational velocity of centre of mass of segment in space in m/s
ω	Absolute angular velocity of segment about a transverse axis through its centre of mass in rad/s
x, y	Horizontal and vertical projections of position in m. single and double dots above indicate first (m/s) and second (m.s^{-2}) time derivatives respectively

Figure C-1. Adapted from Caldwell and Forrester, 1992.

$$\text{TLE} = \sum_{i=1}^3 (m_i g h_i + \frac{1}{2} I_{G_i} \omega_i^2 + \frac{1}{2} m_i v_i^2) \quad (1)$$

Equation C1. Total Limb energy

$$\begin{aligned} \text{PF}_{01} = & \{ \underline{m_i (\ddot{x}_{01} - r_i \omega_i^2 \sin \theta_i + r_i \alpha_i \cos \theta_i) - F_{x02}} \} \dot{x}_{01} \\ & + \{ \underline{m_i (\ddot{y}_{01} + r_i \omega_i^2 \cos \theta_i + r_i \alpha_i \sin \theta_i + g) - F_{y02}} \} \dot{y}_{01} \end{aligned} \quad (2)$$

Equation C2. Joint Power

$$\begin{aligned} \text{PM}_{01} = & \{ \underline{(I_{G_i} + m_i r_i^2) \alpha_i + m_i r_i \ddot{x}_{01} \cos \theta_i} \\ & + \underline{m_i r_i (\ddot{y}_{01} + g) \sin \theta_i} \\ & - (F_{x02} \cos \theta_i + F_{y02} \sin \theta_i) l_i - C_{02} \} \omega_i \end{aligned} \quad (3)$$

Equation C3. Muscle Power

$$\text{PF} = \mathbf{F} \cdot \mathbf{V} = FV \cos \beta = F_x V_x + F_y V_y \quad (4)$$

Equation C4. Joint Power during swing phase

$$\text{IP}_T = \text{PF}_H + \text{PF}_K + \text{PM}_H + \text{PM}_{K(T)} \quad (5)$$

Equation C5. Instantaneous Power of a segment

Equations C1-C5. Adapted from Caldwell and Forrester, 1992.







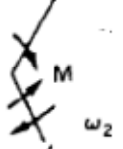

Description of movement	Type of contraction	Directions of segmental ang. velocities	Muscle function	Amount, type and direction of power
Both segments rotating in opposite directions (a) joint angle decreasing	Concentric		Mechanical energy generation	$M\omega_1$ generated to segment 1. $M\omega_2$ generated to segment 2.
(b) joint angle increasing	Eccentric		Mechanical energy absorption	$M\omega_1$ absorbed from segment 1. $M\omega_2$ absorbed from segment 2.
Both segments rotating in same direction (a) joint angle decreasing (e.g. $\omega_1 > \omega_2$)	Concentric		Mechanical energy generation and transfer	$M(\omega_1 - \omega_2)$ generated to segment 1. $M\omega_2$ transferred to segment 1 from 2.
(b) joint angle increasing (e.g. $\omega_2 > \omega_1$)	Eccentric		Mechanical energy absorption and transfer	$M(\omega_2 - \omega_1)$ absorbed from segment 2. $M\omega_1$ transferred to segment 1 from 2.
(c) joint angle constant ($\omega_1 = \omega_2$)	Isometric (dynamic)		Mechanical energy transfer	$M\omega_2$ transferred from segment 2 to 1.
One segment fixed (e.g. segment 1.) (a) joint angle decreasing ($\omega_1 = 0, \omega_2 > 0$)	Concentric		Mechanical energy generation	$M\omega_2$ generated to segment 2.
(b) joint angle increasing ($\omega_1 = 0, \omega_2 > 0$)	Eccentric		Mechanical energy absorption	$M\omega_2$ absorbed from segment 2.
(c) joint angle constant ($\omega_1 = \omega_2 = 0$)	Isometric (static)		No mechanical energy function	Zero

Figure C-2. The mechanisms by which “power” (as incorrectly termed by the authors) is generated, absorbed and transferred at a joint. Really, it is a generation, absorption and transfer of energy at a joint. Adapted from Robertson and Winter, 1980.

APPENDIX D

INFORMED CONSENT AND HEALTH HISTORY FORMS

Consent Form for Participation in a Research Study
University of Massachusetts Amherst

Principal Investigator: **Graham Caldwell PhD., Kinesiology Dept.**
Student Researcher: **Devon Frayne BSc., Kinesiology Dept.**
Study Title: *Kinetic Asymmetries during Submaximal and Maximal Running*

1. WHAT IS THIS FORM?

This form is called a Consent Form. It will give you information about the study so you can make an informed decision about participation in this research study. It will also describe what you will need to do to participate and any known risks, inconveniences or discomforts that you may have while participating. If you decide to participate, you will be asked to sign this form and you will be given a copy for your records.

2. WHO IS ELIGIBLE TO PARTICIPATE?

We will recruit healthy, young female competitive speed and power athletes (i.e. sprinters, football and soccer players) aged 18 to 26 who regularly engage in intense running, and are familiar and comfortable with treadmill running. Subjects will complete a Health Assessment and Demographics Form to ensure that they have no medical conditions (e.g. musculoskeletal injuries within the last year requiring surgery, deformities, etc.) that would prevent participation in exercise of this nature. In addition, these subjects must be able to sprint on the force treadmill at a minimum speed of 8 m/s as determined during the first visit.

3. WHAT IS THE PURPOSE OF THIS STUDY?

The purpose of this research is to investigate asymmetry in kinetic (forces and torques) and kinematic (movement) parameters in subjects while they sprint on a treadmill at speeds ranging from 70% to 100% maximal. This particular treadmill is instrumented with a force platform beneath the belt to obtain instantaneous ground reaction force data. Subjects will be required to perform brief (5 second) bouts of sprinting on the treadmill with ample time to rest and recover in between. Your body motions (i.e. how you move your arms and legs) will be recorded and analyzed to calculate how similar your movements are in your left and right limbs. Results from the sprint bouts will be compared to investigate the amount and type of asymmetry, and how these change as maximal speed is approached. These results will be used to determine if asymmetry could limit maximal sprinting speed, which has implications for coaches, trainers and sport medicine professionals.

4. WHERE WILL THE STUDY TAKE PLACE AND HOW LONG WILL IT LAST?

The study will be conducted in the Biomechanics Laboratory of the Department of Kinesiology, located in the basement of the Totman Building at 30 Eastman Lane, Amherst, MA, on the campus of the University of Massachusetts Amherst. It is expected that each subject will be required to make two visits to the Totman Building, with the option of an additional practice session should the subject choose. On visit one the subject will be introduced to the study, read and complete the Health Assessment and Demographics and informed consent documents. The subjects will also undergo a practice session on the treadmill to familiarize and to ensure that they meet the minimum speed requirement of 8 m/s. This visit will last no more than one hour. On visit two the required motion data will be collected in a data collection session that should last 1 to 2 hours. Prior to the second visit, an optional 1 hour force treadmill practice session will be offered to those subjects who felt as though a single hour of practice on the force treadmill was not enough. It is anticipated that data collection for the entire group of subjects will take place between February and May, 2012. We do not anticipate contacting the participants in the future, except to provide them with results from the study.

5. WHAT WILL I BE ASKED TO DO?

If you agree to take part in this study, you will be asked to make the following two visits to the Biomechanics Laboratory of the Kinesiology Department in the Totman Building, 30 Eastman Lane on the University of Massachusetts Amherst campus.

Visit 1

On Visit 1, you will be screened for health and exercise status. You will be asked to read and complete the Health Assessment and Demographics and Informed Consent documents. Only healthy young female subjects who are competitive sprint and power athletes will be permitted in the study.

Assuming you qualify for the study and agree to the informed consent document, you will be given a practice session (or two if you wish) on the force treadmill. During these practice sessions, you may wish to wear comfortable clothing in which to exercise (shorts and T-shirt). During these sessions you will be instructed on how to correctly mount and dismount the treadmill, as well as how the safety harness system works. The practice session should take about an hour, and you will be given a second hour-long practice session if you desire. In addition, your maximal speed will be determined by increasing the treadmill speed incrementally until you cannot maintain the same speed as the treadmill for four seconds. There will be ample time between trials for rest. The fastest speed at which you can maintain proper sprinting motion (not drifting back on the treadmill) will be your 100% maximal speed condition. All other conditions will be percentages of this speed. You will be excluded from this study if your fastest speed does not equal or exceed 8 m/s.

Visit 2

At the end of Visit 1, you will be scheduled for a second visit (Visit 2), which will again take place in the Biomechanics Laboratory in the Totman Building. In Visit 2 we will collect motion and force data while you sprint on the treadmill

You will be asked to bring and wear comfortable clothing in which to exercise (shorts and T-shirt). Small adhesive reflective markers will be placed on your skin to allow us to track the movement of your legs with a motion capture camera system during the exercise sessions.

You will then be asked to complete eight 5 second sprinting trials on the treadmill; two trials each at 70, 80, 90 and 100% of your maximal running velocity as determined from visit one.

You will be asked to complete the following steps:

1. Stretch, and follow your usual warm up on the treadmill.
2. Once warmed up, you will be asked to perform a 5 second sprint bout at a predetermined speed that represents 70% of your maximal speed. During the exercise, the investigators will use a digital motion capture system (Qualysis) to record the motion of the reflective markers on your legs. A force platform mounted beneath the treadmill belt will measure the forces that you apply on the treadmill. Motion and force records will be stored on computer for subsequent analysis. After the completion of the 70% condition data collection, you will be given a rest period long enough to decrease your heart rate to 100 beats per minute (as measured with a stopwatch and your carotid pulse).
3. After the rest period, the speed of the treadmill will be increased to 80% of your maximum and you will complete another 5 second sprint trial with the same data collected as in the previous condition.
4. Steps 2 and 3 will be repeated for the 90 and 100% maximal speed conditions. Then, the same steps will be repeated in the opposite order from 100% to 70% maximal speed. It is anticipated that the entire testing session will be completed in 1 to 2 hours. You will be undergoing exercise for approximately 40 seconds in total during this entire session, divided into eight 5 second bouts, with opportunity for rest and cool down and recovery between and after the sprints respectively. Therefore, of the estimated two hour duration of this visit, your time spent actually sprinting at high speed will be limited to less than one minute. The rest of the time will consist of the investigator placing reflective markers and calibrating the system, your warm-up, and cool-down/recovery and rest times before, after and in between your 8 short sprint bouts.

6. WHAT ARE MY BENEFITS OF BEING IN THIS STUDY?

You will receive no direct benefit from participating in this study. As an indirect benefit, you may discover information about your sprinting ability. Any information that is obtained from this study will be made available to you upon your request. The purpose of the study is to provide the investigators with information that will aid our future research of asymmetry and sprint performance in humans. Thus, another indirect benefit of participation in this study is the opportunity to contribute to our knowledge in this area.

7. WHAT ARE MY RISKS OF BEING IN THIS STUDY?

As with any exercise, you may become temporarily fatigued during the exercise sessions, which would be normal. Relatively minor muscular discomfort might be experienced during these exercise sessions, but any muscle fatigue and soreness is temporary and complete recovery will occur within 30 minutes of test completion. Some slight muscle soreness may appear after the exercise sessions, but should not interfere with your normal daily activity, and will dissipate over the next 48 hours. All muscle discomfort should be no more than is typically seen with your regular exercise routine as a trained power/speed athlete. Since you will only be sprinting for a total of 40 seconds (separated into individual bouts of ~5 seconds per bout), the demands of this study should be less than those you experience during competition and most of your training sessions. In an attempt to minimize discomfort, participants will be given ample time to stretch and warm up their muscles prior to, and cool down after, the exercise sessions.

Another potential risk is stumbling and falling during one of the sprint trials. To alleviate this risk, you will be supported by a safety harness system with a spotter to ensure that no harm will come to you if you stumble during one of the high speed conditions. This safety harness system has been tested and halts forward falling progress, suspending the subject above the treadmill if balance is lost.

You may find the study procedures inconvenient in that they will take a few hours of your time. You will not be exercising during the entire data collection period, and will get ample time to rest. As with any exercise, you may wish to refrain from eating immediately prior to the exercise session.

8. HOW WILL MY PERSONAL INFORMATION BE PROTECTED?

Procedures will be put in place to protect the confidentiality of your study records. Research records will be labeled with a code that is linked to individual subjects, and all data files will be identified by the code only. The researchers will keep all study records (including any codes to your data) in a secure location (locking file cabinet as an example). A master key that links names and codes will be maintained in a separate and secure location. The master key will be destroyed after the close of the study. Any computer hosting data files will have password protection to prevent access by unauthorized users. Only the members of the research staff for this specific study will have access to the passwords. At the conclusion of this study, the researchers may

publish their findings. Information will be presented in summary format and you or any other individual subject will not be identified in any publications or presentations.

9. WILL I RECEIVE ANY PAYMENT FOR TAKING PART IN THE STUDY?

No payment will be provided for participating in this study.

10. WHAT IF I HAVE QUESTIONS?

Take as long as you like before you make a decision as to whether you wish to participate in this study. We will be happy to answer any questions you have about this study. If you have further questions about this project or if you have a research-related problem, you may contact the principal investigator, Graham Caldwell (545-0017; gc@kin.umass.edu) or the student researcher Devon Frayne (545-6075; dfrayne@kin.umass.edu). If you have any questions concerning your rights as a research subject, you may contact the University of Massachusetts Amherst Human Research Protection Office (HRPO) at (413) 545-3428 or humansubjects@ora.umass.edu.

11. CAN I STOP BEING IN THE STUDY?

You do not have to be in this study if you do not want to. If you agree to be in the study, but later change your mind, you may drop out at any time. There are no penalties or consequences of any kind if you decide that you do not want to participate.

12. WHAT IF I AM INJURED?

The University of Massachusetts does not have a program for compensating subjects for injury or complications related to human subjects research. However, in the unlikely event of an injury, the study personnel will assist you in contacting medical personnel who can provide you treatment.

13. SUBJECT STATEMENT OF VOLUNTARY CONSENT

I have read this form and decided that I will participate in the project described above. The general purposes and details of the study as well as possible hazards and inconveniences have been explained to my satisfaction. I understand that I can withdraw at any time.

Participant Signature

Print Name

Date

By signing below I indicate that the participant has read and, to the best of my knowledge, understands the details contained in this document and has been given a copy.

Signature of Person
Obtaining Consent

Print Name

Date

Health Assessment and Demographics Form

Date _____

Participant Number _____

Please answer the following questions to the best of your knowledge (circle YES or NO)

- 1. YES NO Has a doctor ever said you have a heart condition and recommended only medically supervised activity?
- 2. YES NO Do you ever suffer pains in your chest brought on by physical activity
- 3. YES NO Have you developed chest pain in the last month?
- 4. YES NO Do you ever feel faint or have spells of severe dizziness, passed out, palpitations or rapid heart beat?
- 5. YES NO Has the doctor ever told you that your blood pressure was too high? (systolic \geq 160 mm Hg or diastolic \geq 90 mm Hg on at least 2 separate occasions?)
- 6. YES NO Do you smoke cigarettes?
- 7. YES NO Do you have a bone or joint that could be aggravated by the proposed physical activity?
- 8. YES NO Do you have diabetes?
- 9. YES NO Do you have a family history of coronary or other atherosclerotic disease in parents or siblings prior to age 55?
- 10. YES NO Has your serum cholesterol ever been elevated?
- 11. YES NO Is there any physical reason not mentioned here why you should not participate in high speed running even if you wanted to?

Please provide an explanation below for any of the questions to which you answered YES:

Participant's initials _____

Health Assessment and Demographics Form

Date _____

Participant Number _____

Age (in years) _____

Gender (circle one) M F

Height _____ feet _____ inches or _____ cm

Weight _____ lbs or _____ kg

Please circle one:

Do you regularly engage in intense physical activity (ie. competing in sports)? YES NO

Do you regularly train at high speeds? (ie fast sprints, short durations)? YES NO

Do you currently have any injuries that may affect the way you walk or run? YES NO

If YES, please describe the injury, and when it happened:

Did you injure your lower extremity in the last year? YES NO

If YES, please describe the injury and when it happened:

If you answered YES to the above question, did this injury require surgery? YES NO

Participant's initials _____

APPENDIX E

ABSOLUTE ASYMMETRY SCORES AND PERCENT DIFFERENCES

$$\text{AbsSA} = \frac{\left| 45^\circ - \arctan(X_{\text{left}}/X_{\text{right}}) \right|}{90^\circ} \times 100\% \quad (\text{E.1})$$

Equation (E.1) is used to calculate the absolute asymmetry scores (AbsSA) for each parameter values (X_{left} and X_{right}).

Table E.1: Selected asymmetry score values and percent differences. Percent difference refers to the absolute percentage difference between two values. The results of performing the operation enclosed in brackets from Equation E.1 are depicted in the Quotient columns. AbsSA refers to the absolute asymmetry score (AbsSA in equation E.1) calculated from the corresponding quotients.

Percent Difference	Quotient ($X_{\text{left}} > X_{\text{right}}$)	Quotient ($X_{\text{left}} < X_{\text{right}}$)	AbsSA
1	1.01	0.99	0.32
2	1.02	0.98	0.64
3	1.03	0.97	0.97
4	1.04	0.96	1.3
5	1.05	0.95	1.63
10	1.11	0.9	3.35
15	1.18	0.85	5.15
20	1.25	0.8	7.04
30	1.43	0.7	11.12
40	1.67	0.6	15.6
50	2	0.5	20.48

We reported asymmetries as percent differences in some places in the Discussion section because inter-limb differences are more commonly reported as percent differences (i.e. 5%) as opposed to absolute asymmetry scores (i.e.1.63). Table E.1 is a truncated version of the reference table used by the authors to convert absolute asymmetry scores into percent differences. It is important to note that quotients can be derived for highly precise percentage differences however this truncated version will serve as an appropriate explanatory tool. The reader should begin by looking at the percent difference column which reveals absolute differences between limbs for a given parameter. A difference between limbs arises from two possible scenarios; either the left limb's value exceeds the right one, or the converse. The second column from the left reports the results of dividing X_{left} by X_{right} when the left limb value is greater than the right. The third column from the left reports the results of dividing X_{left} by X_{right} when the left limb value is smaller than the right. The reader should note that there is only one absolute asymmetry score that corresponds to a percentage difference, even though the percentage difference can arise from two possible conditions. The non-linear nature of AbsSA scores is also shown in the Methods in Figure 3.2.

APPENDIX F

FURTHER STATISTICAL EXPLANATION

Three main effects (subject differences, speed and trial) were examined in an ANOVA procedure for each key sprinting parameter, with speed being the effect of interest. The effects of speed were examined using the subject*speed mean square as the error term. To determine if the order in which the trials were completed had an effect on our results, the interaction of speed and trial was examined using the subject*speed*trial mean square as the error term.

Table F-1: Significance (p-value) for the main effect of speed and the speed*trial interaction effect. Parameters for which speed or speed*trial interaction significantly affected asymmetry are denoted by (*).

Parameter	p-value (Speed)	p-value (Speed*Trials)
A/P Braking Impulse*	0.03	0.52
Flight Time	0.55	0.25
Peak Eccentric Knee Flexion	0.74	0.56
Step Length	0.7	0.18
Stride Length	0.91	0.38
Peak Concentric Hip Flexion	0.78	0.6
Stance Time	0.58	0.38
Step Time	0.7	0.18
Peak Eccentric Knee Flexion Timing	0.25	0.65
Vertical Propulsive Impulse	0.32	0.47
A/P Propulsive Impulse	0.25	0.79
Effective Vertical Stiffness*	0.0006	0.0512

If the effect of speed on asymmetry was significant (A/P Brake Impulse and Effective Vertical Stiffness), orthogonal polynomial contrasts were employed to determine the nature of this effect.

Table F-2: Tests of significance for first (linear), second (quadratic) and third (cubic) order relationships.

Parameter	Contrast p-value		
	Linear	Quadratic	Cubic
A/P Braking Impulse	0.08	0.21	0.03
Effective Vertical Stiffness	<0.05	0.72	0.42

Linear and cubic functions significantly describe the relationships between speed and asymmetry for effective vertical stiffness and A/P braking impulse respectively.

APPENDIX G

ASYMMETRY SCORES AS PERCENT DIFFERENCES

Table G-1: Average Absolute Asymmetry Scores (AbSA) and percent differences between limbs (% Diff) for all tested temporal parameters (Mean ± SD) across all speed conditions.

ASYMMETRY SCORE MEAN (± SD)	70		80		90		100	
	AbSA	% Diff	AbSA	% Diff	AbSA	% Diff	AbSA	% Diff
Stride Length	0.41 (0.35)	~1%	0.37 (0.39)	~1%	0.40 (0.37)	~1%	0.44 (0.65)	~1%
Step Length	0.85 (1.03)	~3%	0.69 (1.00)	~2%	0.84 (1.05)	~3%	0.68 (0.58)	~2%
Step Time	0.85 (1.03)	~3%	0.69 (1.00)	~2%	0.84 (1.05)	~3%	0.68 (0.58)	~2%
Flight Time	4.15 (5.10)	~12%	2.77 (3.79)	~8%	3.25 (3.76)	~10%	3.30 (4.38)	~10%
Stance Time	1.23 (1.88)	~4%	0.82 (1.10)	~3%	1.18 (1.72)	~4%	0.91 (0.89)	~3%
Peak Knee Flexion Power Timing	0.77 (0.52)	~2%	0.86 (1.93)	~3%	1.00 (3.20)	~3%	0.74 (0.76)	~2%
Peak Hip Flexion Power	4.55 (2.81)	~13%	5.41 (7.26)	~15%	4.65 (5.45)	~13%	3.68 (4.93)	~11%
Peak Knee Flexion Power	4.33 (4.33)	~13%	3.88 (3.41)	~11%	4.5 (4.77)	~13%	3.73 (4.11)	~11%
Vertical Propulsive Impulse	7.99 (15.16)	~22%	9.22 (16.51)	~25%	14.49 (20.4)	~37%	12.72 (18.88)	~34%
A/P Propulsive Impulse	8.21 (14.99)	~23%	9.98 (16.36)	~28%	13.44 (19.23)	~35%	15.09 (24.79)	~39%
A/P Brake Impulse	5.17 (3.85)	~15%	6.06 (4.28)	~17%	4.81 (4.2)	~13%	7.14 (7.26)	~20%
Effective Vertical Stiffness	3.87 (2.72)	~11%	5.88 (4.34)	~17%	6.54 (4.97)	~19%	8.11 (6.06)	~22%

% Diff = 100 - [(Smaller Value/Larger Value) * 100]

i.e. Left limb value= 15, Right limb value=20

% Diff= 100 - [(Left/Right) * 100]
= 25

The left limb's value is 75% of that of the right limb. So the left limb's value is 25% lower than the right.

APPENDIX H

EXEMPLAR KINEMATIC AND KINETIC DATA

This appendix contains data from an exemplar subject (Subject 6) to describe general kinematic and kinetic patterns of our subjects. For all graphs the 70%, 80%, 90% and 100% Maximal Speed conditions are denoted by thin-solid, thin-dotted, thin-dashed and thick-dotted lines respectively.

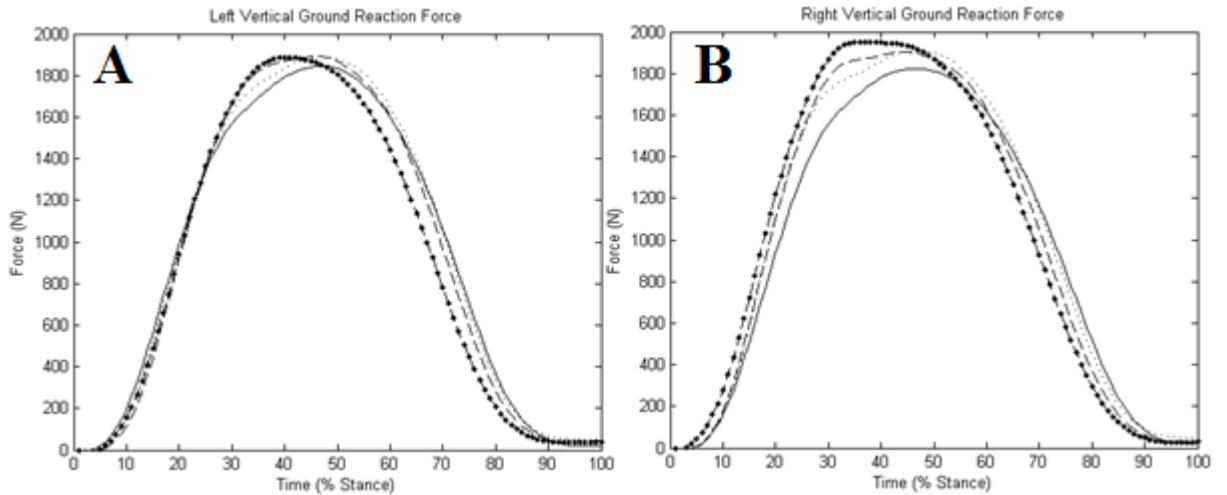


Figure H.1. Average vertical ground reaction profiles.

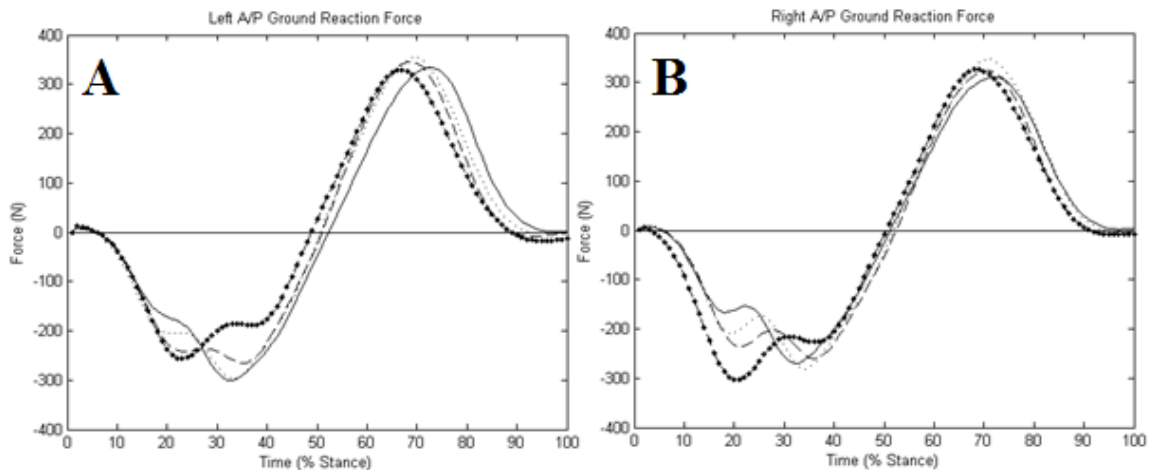


Figure H.2. Average Antero-posterior (A/P) ground reaction profiles. Negative and positive values indicate braking and propulsion respectively.

The following graphs are of joint angles, angular velocities and net joint moments across a single left stride. For the graphs of joint angle, positive values indicate a relatively dorsi-flexed ankle, an extended knee and flexed hip compared to anatomical position. For the graphs of joint angular velocity and net joint moment, positive values indicate ankle dorsi-flexion, knee extension and hip flexion. Negative values represent the opposite orientation, joint motion or net joint moment. In all cases, the vertical dashed line indicates toe-off.

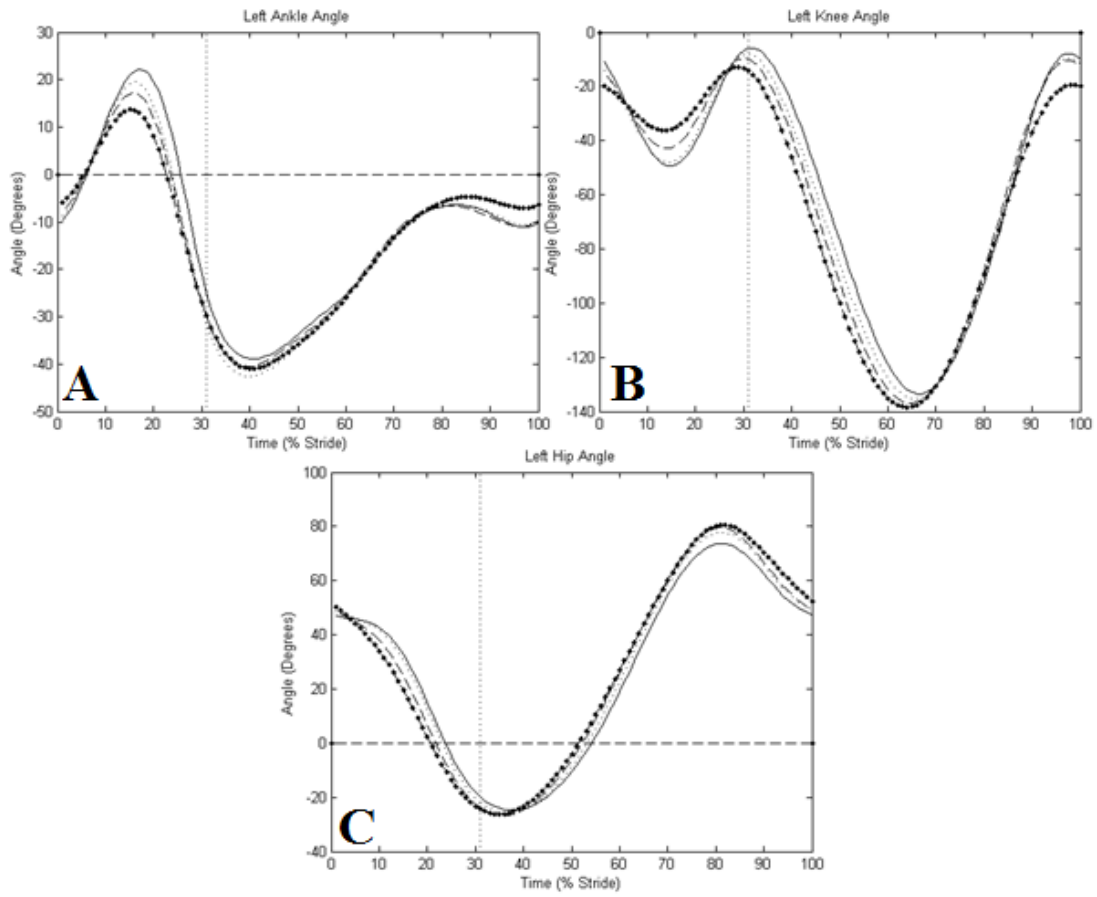


Figure H.3. Left Ankle (A), Knee (B) and Hip (C) sagittal plane joint angles.

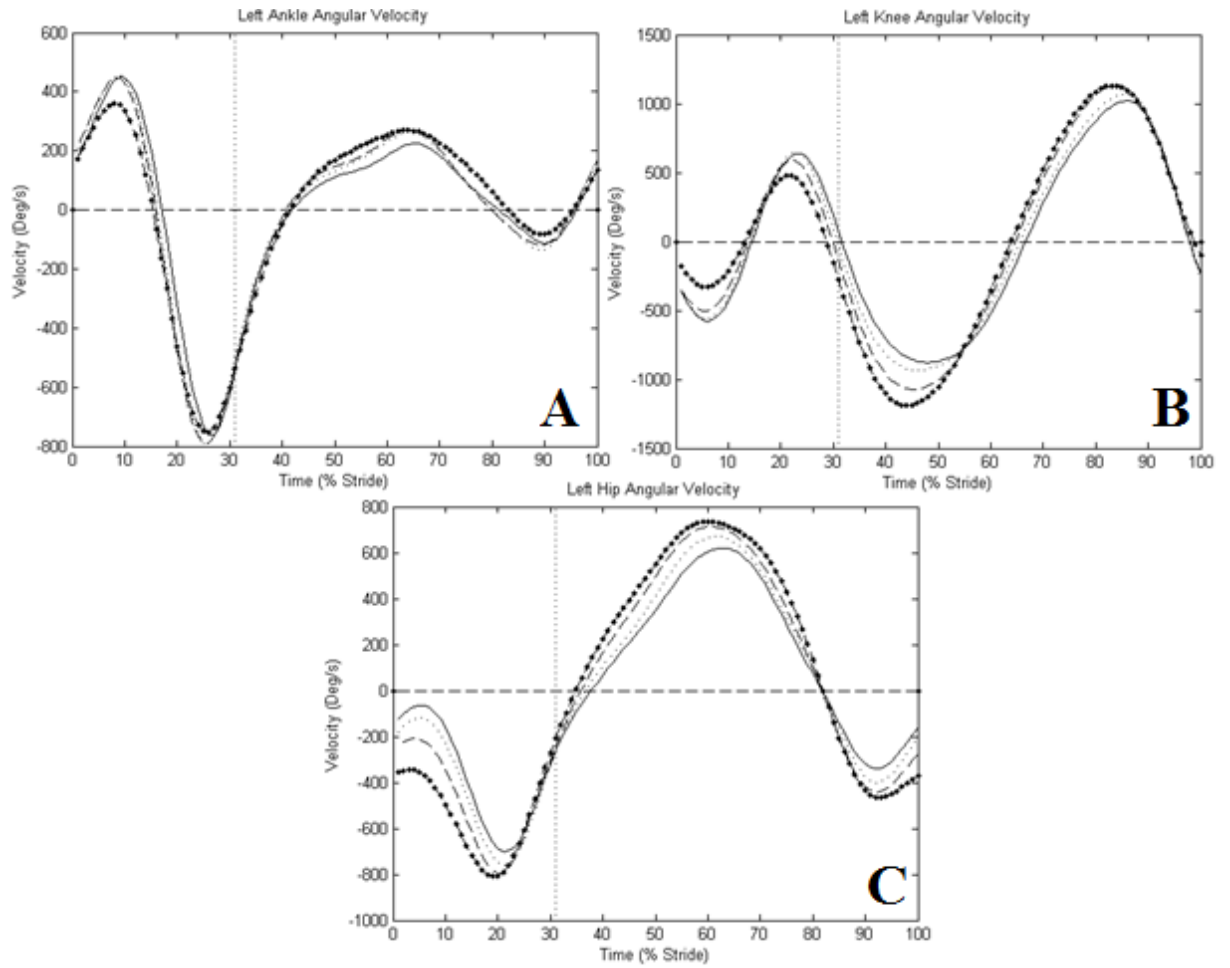


Figure H.4. Left Ankle (A), Knee (B) and Hip (C)sagittal plane joint angular velocities.

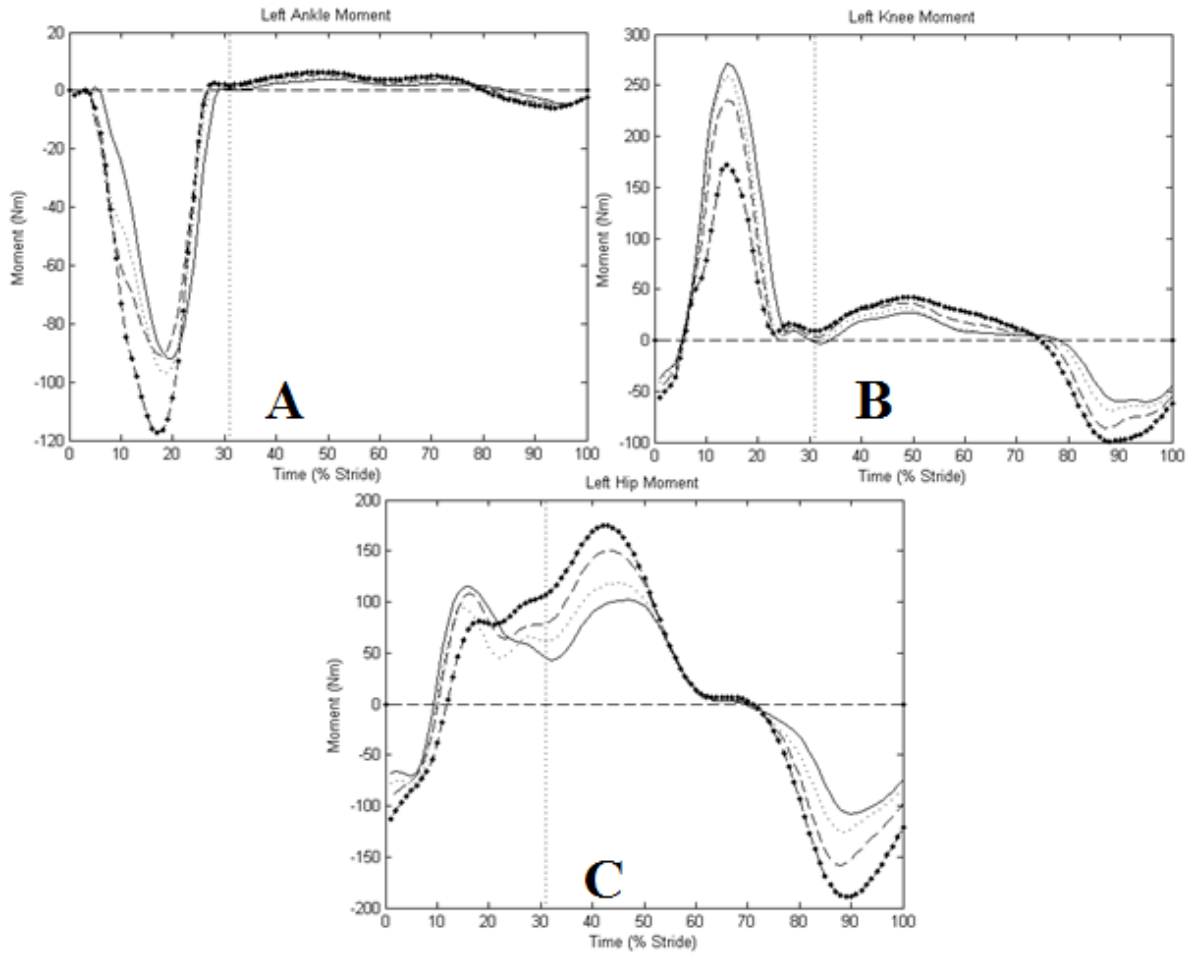


Figure H.5. Left Ankle (A), Knee (B) and Hip (C) sagittal plane net joint moments.

The final set of graphs depicts muscle moment powers and segmental energy profiles during a left stride. Positive and negative muscle moment powers are associated with concentric (energy generation) and eccentric (energy absorption) muscle action across the joint respectively. Again, the vertical dashed line demarcates toe off.

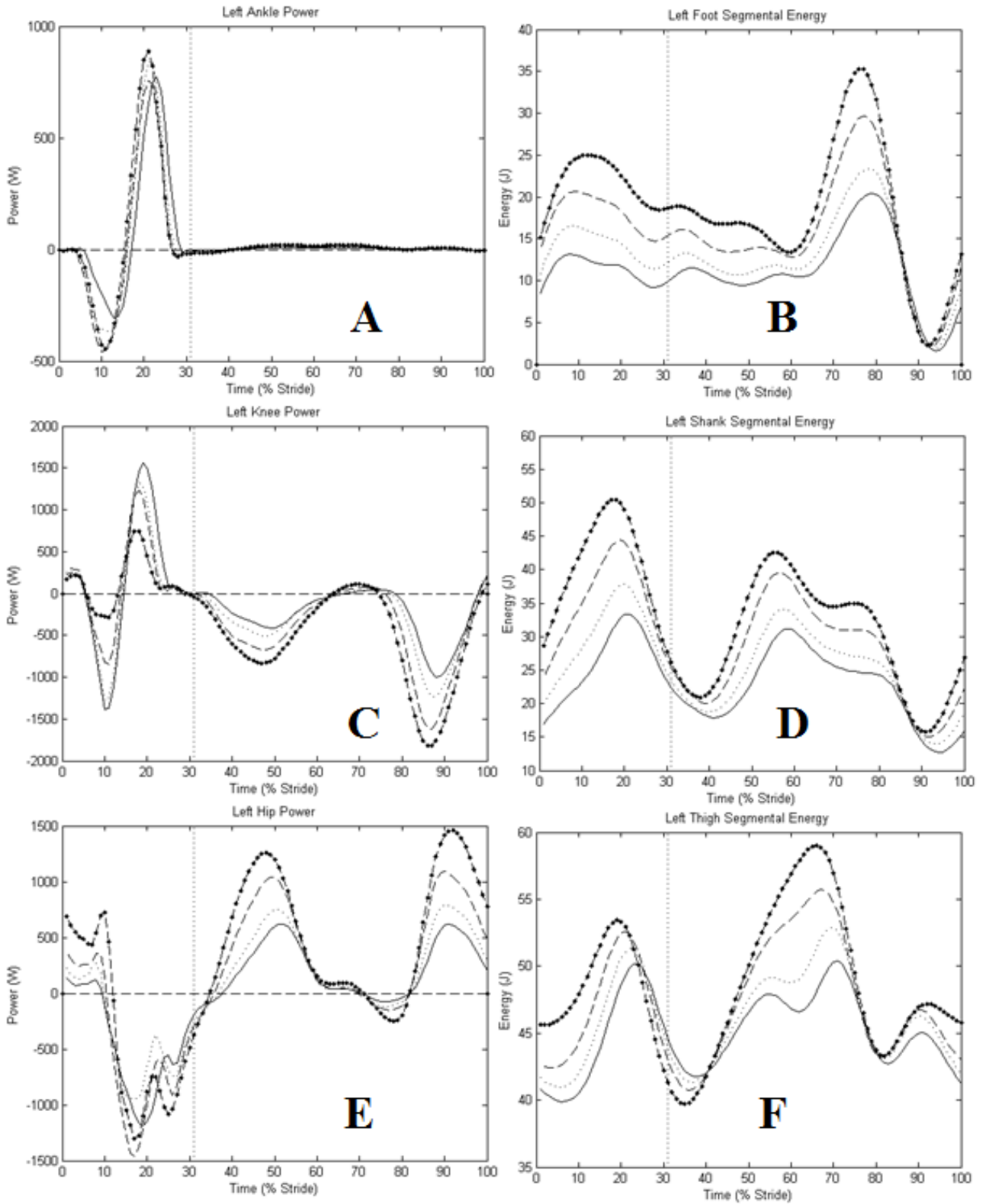


Figure H.6. Left Ankle (A), Knee (C) and Hip (E) sagittal plane net muscle moment powers, and Foot (B), Shank (D) and Thigh (F) segmental energy profiles.

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