University of Alberta

Analysis and Simulation of the Recovery Leg During Sprinting

by

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ABSTRACT

In this investigation of the recovery leg during sprinting, two projects were completed. The objective of the first project was to examine the relationships among the kinematic and kinetic variables associated with the recovery phase of sprinting in order to better understand the differences seen among individual sprinters. High speed video was collected during competition of 14 elite male 100 metre sprinters. Data reduction and analysis was completed using the Ariel Performance Analysis System (APAS), and an inverse dynamics approach was used to calculate the resultant joint moments at the hip, knee, and ankle for one complete sprint stride. Pearson product-moment correlation coefficients were calculated to determine where relationships existed among the kinematic and kinetic variables which interact to govern the movement of the recovery leg during sprinting. The objective of the second project was to use computer simulation to investigate the effect changing hip and knee angles at takeoff would have on the subsequent kinematics of the recovery leg. A planar, four segment model was created using Working Model 2D to simulate the leg during the recovery phase of sprinting, and included a thigh, shank, foot and body segment. The model was developed based on data from the 14 elite sprinters from the previous project. Anthropometric characteristics were used to construct the model, and hip, knee, and ankle resultant joint moments were input to drive the simulation. The simulation was run with modified initial conditions, in which the hip and knee extension angles were systematically increased or decreased. The results showed that there are a number of statistically significant correlations

among the kinematic and kinetic variables associated with the recovery leg. These correlations were seen throughout the recovery phase, from takeoff to touchdown. In addition, the computer simulation showed that the kinematics of the recovery leg during sprinting can be altered by modifying the hip and knee positions at takeoff. Changes in hip angle were found to be more influential on the overall leg movement than changes at the knee. The results from these studies have important implications for sprint training and speed development.

DEDICATION

To Kim, for her love and support

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CHAPTER 1

Introduction

Biomechanics is defined as the science that examines the internal and external forces acting on a body, and the effects produced by these forces (Hay, 1993). Biomechanical considerations in human movement are evident in a number of disciplines, including biomedical engineering, ergonomics, and orthopaedics, to name a few. When biomechanics is applied to sport, one of the objectives is to understand the technique used by athletes performing a specific movement or skill and provide answers to questions such as "what is the best foot position for the sprint start?" or "why is it beneficial to use the hitch-kick in long jumping?" To complete this type of research requires the application of experimental methods.

Experimental studies, in which the investigator is in control of the testing environment, can provide a greater understanding of the technique used for a given movement. For example, Schot and Knutzen (1992) examined the biomechanics of the sprint start using four different start positions, and identified how changing the arm orientation and block distance influence critical aspects of the sprint start. From this experimental research, direct intervention can then be used to modify an athlete's technique in order to improve performance. However, the power of such experimental studies can be limited in its application, as making modifications in one aspect of technique may inadvertently introduce changes to another (Yeadon & Challis, 1994).

Two basic approaches may be taken to complete experimental studies in sport biomechanics: kinematic and kinetic. Kinematic or time-dependent studies involve quantifying the movement without reference to the forces producing the movement. This type of analysis produces a description of the motion, but it does not provide insight as to what causes a body to move how it does (Hay, 1993). Kinetic studies are those in which the forces causing movement or changes in movement are investigated. Kinetic analyses usually involve some form of modeling of the human as a linked system of rigid bodies. Conceptually, there are two methods to perform kinetic analyses. One is classified as the direct dynamics problems in which the external forces applied to a mechanical system are known and the objective is to determine the motion of the system which results from the applied forces and moments. The other type is defined as the inverse dynamics problems. Here, the motion of the mechanical system is known in various forms but the externally applied forces and moments are to be determined. If the exact motion histories of the system, especially the accelerations, are available, then this type of problem can be solved (Chao & Rim, 1973).

Experimental studies are often used to identify various aspects of skills which are associated with better performances. However, there are limitations to these types of studies in the amount of information and understanding they can give regarding the performance as they are typically based only on a few examples of the movement from a small number of subjects. In addition, such studies do not allow the investigator to quantify the effect of varying technique outside a range of observed performances of a movement, and are thereby limited in their ability to speculate on what is the best technique for a particular movement and athlete. In such circumstances theoretical methods must be applied.

Theoretical studies in biomechanics involve the development of models to simulate the skill mathematically, thereby allowing complete control and flexibility over the technique used for the performance of dynamic human movements. Modeling is appropriate when the system being investigated can be represented physically or mathematically so that a given behaviour can be examined throughout the range of carefully controlled conditions (Miller, 1979). Modeling is also required when there is a chance that a physical test could be injurious to the participant. Models used in biomechanics research have ranged in complexity from simple models (e.g., Farley and Gonzalez, 1996) to very complex models (e.g., Hatze (1981a)). Simulation models have improved the general understanding of the mechanical principles governing human locomotion.

1.1 Sprinting

Sprinting is a basic form of human locomotion, similar to walking or running, in that it is a cyclical pattern of movement that is repeated with each stride. What makes sprinting different from the other means of locomotion is that it is intended to minimize the time of movement, to move from one place to another as fast as possible. It is this factor which makes sprinting an important element in understanding human movement, as well as in improving athletic potential. It is therefore significant that running is not fully understood, particularly from a biomechanical perspective (Vaughan, 1983a).

There is an extensive amount of literature on the biomechanics of sprinting; however there is limited work on the recovery phase of the sprint stride. Once the foot leaves the ground after propulsion, the entire leg must be moved from a rear position to a forward position at a rate exceeding that of the body in time for the subsequent contact phase. The importance of this movement is best supported by the fact that if the recovery leg does not swing sufficiently forward in front of the body the runner would fall to the ground (Dillman, 1971).

The recovery phase has been shown to be the rate limiting factor during sprinting (Chapman & Caldwell, 1983), therefore good recovery mechanics should be the goal of all sprinters, particularly those at the elite level. This includes a lack of full knee extension at takeoff (Mann, 1985), good knee flexion and a high knee lift (Sinning & Forsyth, 1970), a rapid foot descent with the foot moving backwards relative to the body as it strikes the ground (Mann, 1985), and contact made at the point almost beneath the body's centre of gravity (Kunz & Kaufman, 1981; Mann, 1985). Results from studies involving elite sprinters (Mann & Herman, 1985; Kersting, 1999; Kivi, 1999) revealed that sprinters do not all recover the leg in the same manner, with differences seen in a number of variables throughout the entire recovery phase. These differences, however, have only been described by their kinematics; there has been no research completed to explain them from a kinetic standpoint. In order to understand more fully the differences among sprinters, it is also necessary to consider the kinetics of motion because the movement of the leg during the recovery phase of sprinting is governed by the interaction among a number of variables, both kinetic and kinematic (Mena, Mansour, & Simon, 1981). These include the position of the leg at takeoff, the angular displacements and angular velocities of the limb segments, and the muscle moments controlling the motion at each joint. The way in which these variables interact has not been investigated. If sprinters do not all recover the leg in the same manner, there must be some systematic way that these kinematic and kinetic factors work together to produce the resulting movement pattern.

If a number of variables interact to produce the motion of the recovery leg, it is reasonable that the kinematics of the leg at any point is dictated by preceding positions and movements. For a given action or skill, however, the motion can vary greatly depending on the initial position of the body (Chou, Song, & Draganich, 1995). This means that the position of the leg at takeoff during sprinting may be important in determining the overall movement pattern seen during recovery.

Better sprinters tend to minimize the amount of extension at both the hip and knee at the point of takeoff in order to minimize ground contact time (Mann, 1985), therefore it may be advantageous for an athlete to make modifications in technique in order to improve performance. In order to determine the influence of changing hip and knee angles at takeoff it would be necessary to have a subject modify his/her sprinting mechanics and then quantify the resulting changes in the recovery leg. Performing a study of this nature would likely result in injury to the participant (Vaughan, 1984), and therefore must be completed using computer simulation methods.

1.2 Purpose

There were two objectives to this research project:

(a) To determine the relationships among the kinematic and kinetic variables which govern the movement during the recovery phase of elite sprinters.

(b) To use computer simulation to determine what effect changing hip and knee angles at takeoff would have on the subsequent kinematics of the recovery leg.

To fulfill these objectives, two projects were completed. Project 1 was a kinematic and kinetic analysis of the recovery leg of elite sprinters. Subjects for this project were the semi-finalists and finalists from the men's 100 metres at the 8^{th} IAAF World Championships in Athletics and the 2001 Canadian Senior Track and Field Championships. Statistical methods were used to determine if there were statistically significant correlations among specific kinematic and kinetic variables of the recovery phase within this group of athletes. In Project 2, a planar link segment model for the recovery phase was developed for maximum speed sprinting using computer simulation. Hip and knee angles at takeoff were then modified and the resulting motions were documented.

Using computer simulation to investigate the recovery phase of sprinting was appropriate in this investigation for two reasons. First, the lower extremity movement during sprinting may be assumed to be planar, thereby allowing for a simple yet accurate two-dimensional representation of the dynamical system.

Second, using computer simulation to examine changes to the sprint stride would avoid the risk of injury to the athlete if experimental methods were employed.

This research has a number of direct applications. For the coaching community, the kinematic and kinetic sprint analysis will provide a greater understanding of the techniques of elite sprinters. This information can then be used to develop specific training methods which are suitable for the stride characteristics of each individual. The modeling and simulation will provide information to elite sprinters as to how they may become faster and more efficient through the recovery phase. This research is also relevant to scientists, as it will further the knowledge and understanding of how the human body functions during high velocity movements, and will provide another example of the application of computer simulation to dynamical human movement.

1.3 Limitations

The limitations of this study were seen in the body segment parameter calculations, and in the data collection, reduction, and analysis procedures used. Inertial parameters were calculated based on the work of Zatsiorsky and Seluyanov (1983) as modified by DeLeva (1996), who used radiation techniques to determine segment masses, mass centres, and principal moments of inertia about anteroposterior, transverse, and longitudinal axes in 100 adult males, many of whom were physical education students. The anthropometric characteristics of these subjects may not be correlated highly with elite sprinters but these data are the most appropriate that are presently available as they are based on a large sample of athletic male subjects. The video data collected may have contained errors associated with the physical imperfections in the lens optics and in the misalignment of the cameras. In addition, random error may also exist which may be attributed to the researcher's inability to locate defined joint centres during the digitizing process. These errors were controlled for as much as possible by strict adherence to biomechanical video data collection protocol. Data smoothing was also used to attenuate the experimental error.

1.4 Delimitations

The study was delimited to a two-dimensional examination of the recovery leg during sprinting. Fourteen subjects were selected for investigation and were semi-finalists from the men's 100 metres at the 8^{th} IAAF World Championships in Athletics and at the 2001 Canadian Senior Track and Field Championships. The video rate was 120 Hz.

CHAPTER 2

Review of Literature

2.0 Overview

This chapter is a review of the relevant literature for the analysis and simulation of sprinting. Included are the kinematics and kinetics of sprinting during maximum speed, as well as a discussion of the recovery phase of elite sprinters. This will be followed by a summary of mathematical modeling and computer simulation procedures in sport biomechanics. In addition, topics relevant to biomechanical research including anthropometrics, image analysis, and data smoothing techniques are included in Appendix 1.

2.1 Phases of the Sprint Stride

The sprint stride may be divided into a number of phases, according to their timing and function. The two main phases are the contact phase when the runner is touching the ground, and the flight phase when the runner is not in contact with the ground. These phases are cyclic, repeating with each stride (Schmolinski, 1996; Hay, 1993).

The first phase, contact, may be further divided into three sub-phases. The first sub-phase is the resistive phase, which occurs when the foot contacts the ground in front of the centre of mass (Figure 2-1, photo 1). Initial contact is on the outer edge of the sole, high on the ball (metatarsal-phalangeal joint) of the foot, and moves towards the inside as the whole ball of the foot makes contact under the weight of the body. The initial horizontal ground reaction force is a braking force which acts to slow the sprinter down. In most elite sprinters this phase is very short in duration with the foot landing very close to the line of the centre of mass, which is the imaginary vertical line running downwards from the centre of mass.

The second sub-phase is the support phase, at which time the centre of mass passes over the base of support (Figure 2-1, photo 8). The centre of mass reaches its lowest point during this phase as support flexion at the hip, knee, and ankle decrease the downward velocity to zero and cushion the force of the impact. All of the weight of the sprinter is balanced on the ball of the foot; the heel does not touch the ground. Preventing the heel from contacting the ground decreases the vertical displacement of the centre of mass, enabling the body to move faster through the support phase and into the propulsive phase. The distance between the ground and heel varies in individuals from a few millimetres to 3 or 4 centimetres (Schmolinski, 1996).

The third sub-phase during contact is the propulsive phase, where the body has passed over the centre of mass, and the powerful hip extensor muscles act to propel the runner forwards and upwards (Figure 2-1, photo 3). Schmolinski (1996) considers this phase the most important of the sprint stride. The velocity of propulsion depends mainly on the magnitude and direction of the push-off force. Earlier studies (Bunn, 1978; Dillman, 1975) have indicated that "good" runners fully and rapidly extend the thigh about the hip through the propulsive phase, while "poor" runners do not obtain full extension of the knee until after the foot has left the ground. Mann (1985), Mann and Herman (1985), Tupa, Dzhalilov, and Shuvalov (1991) have disputed this action at the knee,

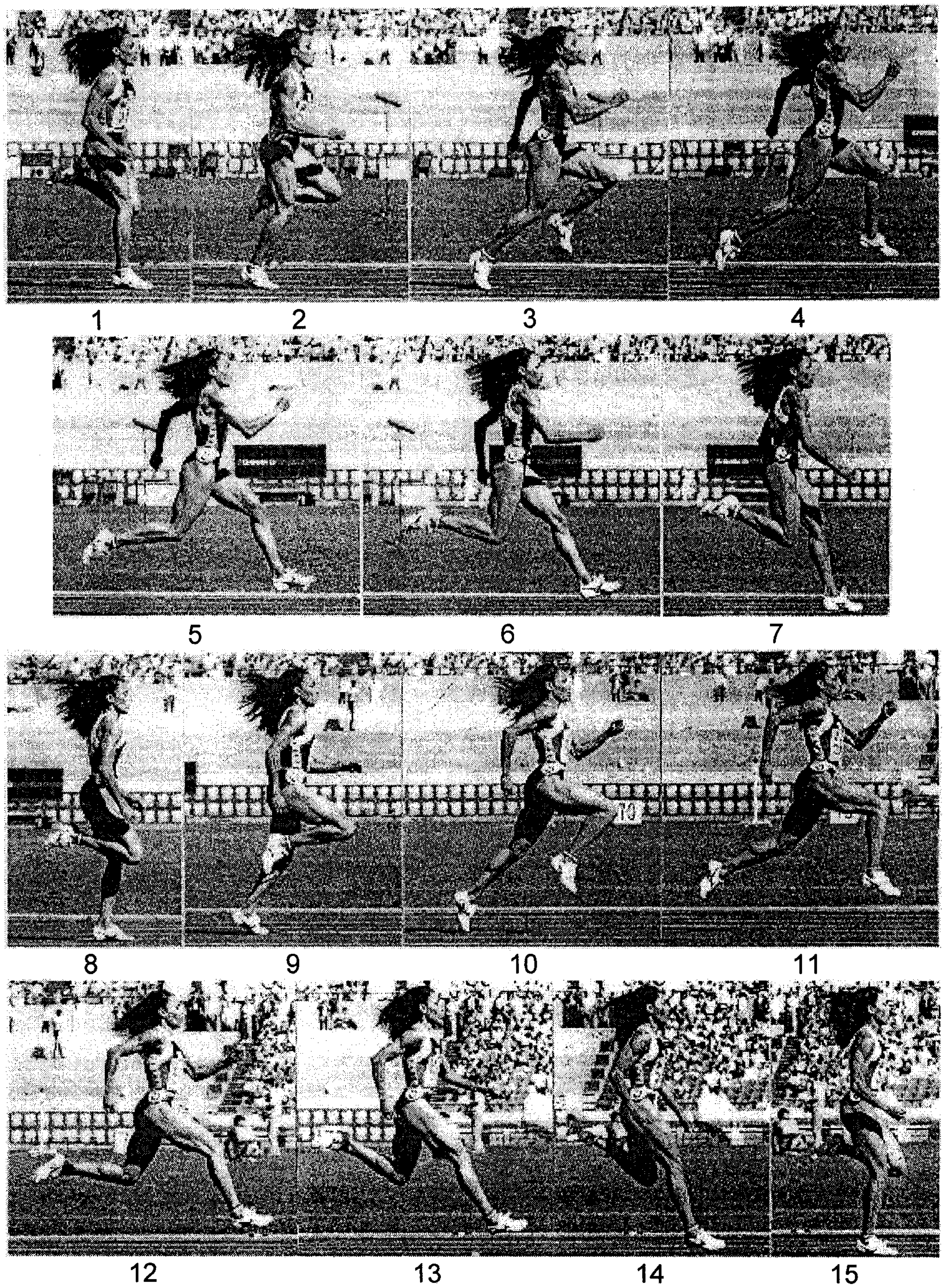


Figure 2-1. Photosequence of elite sprint technique. (Hommel, 1991, pp. 74-75).

stating that better sprinters do not fully extend the knee at takeoff which functions to reduce ground contact time and increase stride frequency.

The flight phase follows the contact phase and the athlete is airborne, during which the centre of mass follows a parabolic pathway (Figure 2-1, photo 4). The centre of mass reaches its highest point midway through this phase. Here, the recovery leg moves from a position behind to in front of the body in preparation for the next ground contact due to powerful contraction of the hip flexors. Leg recovery is best accomplished by flexion of the knee and dorsiflexion of the ankle (Figure 2-1, photo 8). The smallest knee angle should take place at the moment when the knee is positioned vertically below the hip, which will enable the thigh to rotate forward and upward at maximum angular velocity (Schmolinski, 1996) by minimizing the rotational inertia of the swing leg. As the hip approaches maximum flexion, eccentric contraction of the hip extensor muscles slows this forward rotation and the lower leg extends in a relaxed movement (Figure 2-1, photo 10). The hip then begins to extend due to powerful concentric contraction of the hip extensor muscles (Figure 2-1, photo 13). With this action, the athlete attempts to minimize the horizontal velocity of the foot relative to the ground at the instant of ground contact, which would reduce the braking force on contact (Hay, 1993). No sprinter has been able to recover the foot so it is moving backwards relative to the ground at the moment of contact (Mann, 1985), which indicates a braking force is always seen at the moment of ground contact slowing the sprinter down.

2.2 Kinematics of Sprinting

The term "kinematics" refers to the branch of biomechanics that describes how a body moves in space, without reference to the causes of the observed motion (Robertson & Sprigings, 1987). A considerable amount of literature is available describing the kinematics of sprinting.

2.2.1 Horizontal Velocity

Generating a high horizontal velocity of the centre of mass is the key to successful sprinting, as it is the athlete who can produce a high velocity and maintain it through the duration of a race who will win. Northrip, Logan, and McKinney (1974) stated that the theoretical maximal horizontal velocity for humans during running is 12.9 m/s. Horizontal velocities ranging from 8.85 m/s to 10.78 m/s have been previously reported (Armstrong, Costill, & Gehlsen, 1984; Luhtanen & Komi, 1978; Mann & Sprague, 1983; Mann & Herman, 1985), however these studies did not involve elite level sprinters. More recently, Hoskisson and Korchemny (1991) examined elite junior sprinters using high speed cinematography and found a maximal horizontal velocity of 11.9 m/s. Similarly, a study of the men's 100 metre final at the World Championships in Athletics in 1991 reported that the winner of the race achieved a maximal horizontal velocity of 12.05 m/s (Ae, Ito, & Suzuki, 1992).

2.2.2 Vertical Displacement and Velocity

Vertical velocity in sprinting has been virtually ignored in scientific research on sprinting, as it is the component of the resultant velocity which should be optimized. Mann (1985) found that "good" male sprinters attain a mean vertical velocity at takeoff of 0.52 m/s, while "average" male sprinters reach 0.61 m/s, and "poor" male sprinters 0.69 m/s. Classifying sprinters in this manner may provide a general estimate of sprinting ability, but categorizing a sprinter based on vertical velocity may not accurately represent their performance potential. It is possible for a sprinter to have a high horizontal velocity while having a vertical velocity that is larger than ideal. The resultant velocity may be the more important parameter to measure as it would take into consideration both the horizontal and vertical components of the velocity. Mero, Luhtanen and Komi (1986) reported vertical velocities of 0.69 m/s in a group of male sprinters (n = 11, mean 100m time = 10.84 sec) and 0.62 m/s for female sprinters (n = 7, mean 100m time = 11.95 sec). Based on Mann's classification, the males would be considered "poor" and the females "average" sprinters, which is hardly plausible based on each group's mean 100 metre time. These values, in particular the vertical velocity for the males, are higher than ideal. Vertical velocity should be decreased to ensure that the horizontal component of the velocity is maximized

The key to top sprinting is maximizing horizontal velocity, however, some vertical velocity and displacement is necessary in order to provide adequate time for the recovery leg to swing forward and prepare for the next ground contact. During sprinting, the vertical displacement of the centre of mass has been reported at 5.0 cm by Mero, Luhtanen and Komi (1986). A somewhat larger vertical displacement value of 6.7 cm was found by Luhtanen and Komi (1978), but these values may not be comparable as the participants in the study were athletes from different track and field disciplines, including sprinters, throwers, decathletes, and jumpers.

2.2.3 Stride Length and Stride Frequency

Average running speed is the product of stride length and stride frequency:

Average Running Speed (m/s) = Stride Length (m) x Stride Frequency (strides/s)

A stride is identified as the termination of contact of one foot with the ground through the next contact with the same foot, and involves two steps (Adrian & Cooper, 1989). A step is that part of the running stride which begins at the moment when one foot terminates contact with the ground and continues until the opposite foot contacts the surface (Adrian & Cooper, 1989). The term "step" is often used interchangeably with "stride," which is incorrect as they refer to different portions of the sprinting stride.

Stride frequency is the number of strides per second, and is calculated by measuring stride time. Stride time is described as the sum of the time the athlete is in contact with the ground (the contact time), and the time during which the athlete is in the air (the flight time) (Hay & Reid, 1988). Stride frequency is the inverse of stride time. This means if one stride is completed in half a second, then stride frequency is two strides per second. Fast sprinting is achieved by combining a long stride length and a high stride frequency, all other things being equal. If a short-legged sprinter wants to achieve a fast running speed, it would be necessary for this runner to take more strides per unit time than a long-legged sprinter, whose strides are usually longer.

At submaximal running speeds, initial increases in speed are a result of longer stride lengths (Luhtanen & Komi, 1978). As speeds approach maximum, stride length increases level off and stride rate increases (Figure 2-2). Hunter, Marshall, and McNair (2004) stated that each individual's leg length, height of

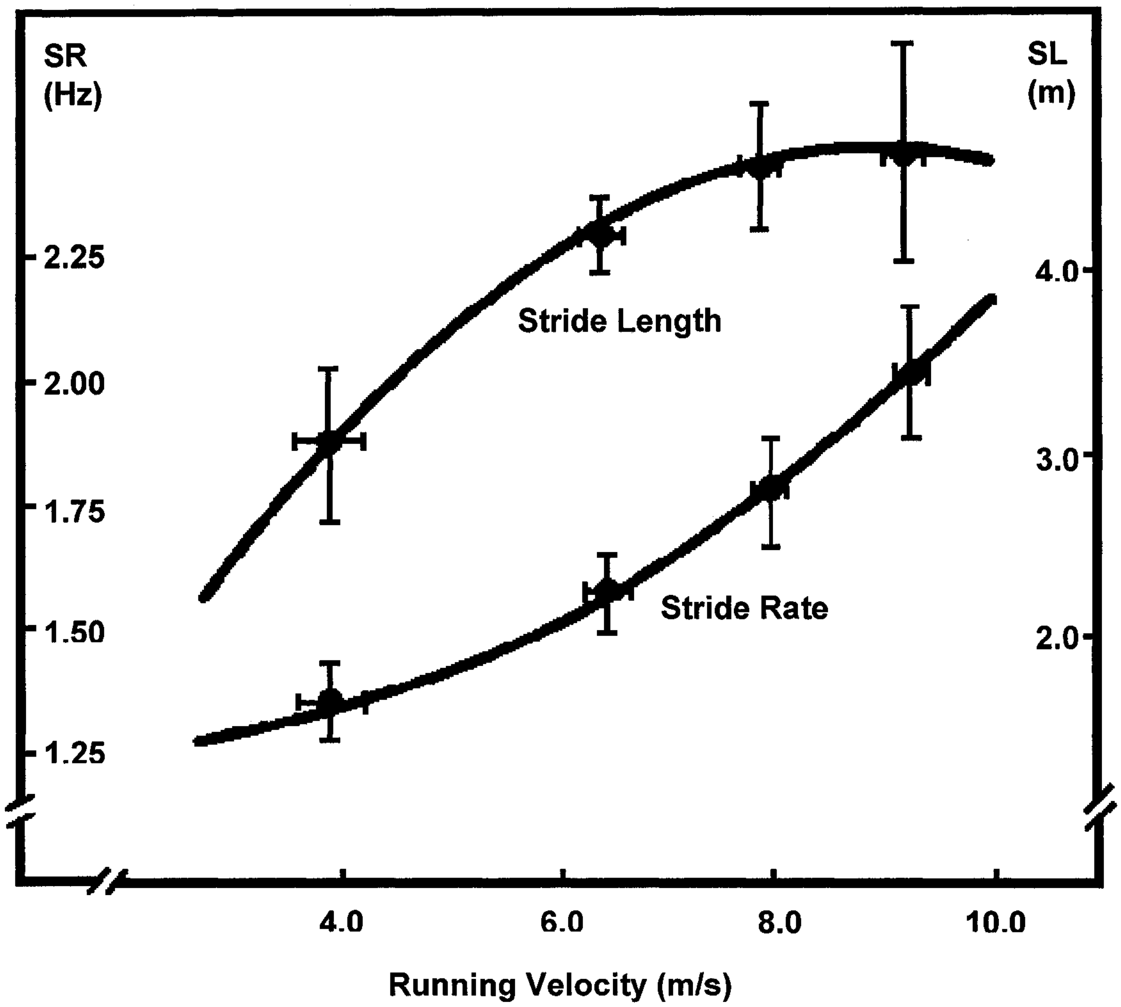


Figure 2-2. Stride length and stride rate when measured at various running velocities (adapted from Luhtanen & Komi, 1978, p. 25.)

takeoff, and vertical velocity of takeoff are possible sources of the interaction between stride length and stride rate. Hay (1993) described step length as the sum of three separate distances: (1) the take-off distance, which is the horizontal distance from the centre of mass to the toe of the take-off foot at the instant it leaves the ground, (2) the flight distance, which is the horizontal distance that the centre of mass travels during the flight phase, and (3) the landing distance, which is the horizontal distance from the centre of mass to the toe of the foot at the instant of ground contact (Figure 2-3).

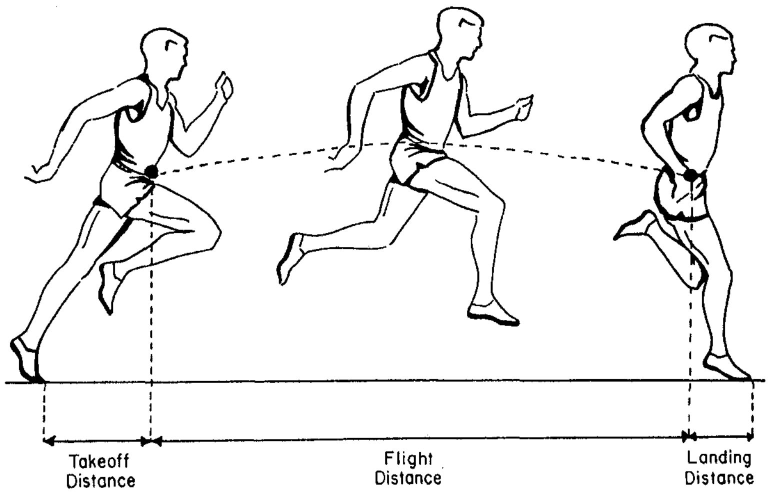


Figure 2-3. Contributions to step length in sprinting (Hay, 1993, p. 398).

The average maximum stride length of top male sprinters has been reported as 4.50 m (Nummela, Vuorimaa, & Rusko, 1992) in national level sprinters, while Hoskisson and Korchemny (1991) have found values in elite junior sprinters ranging from 4.50 to 4.72 m. At the 1991 World Championships in Athletics, maximum stride lengths for the eight finalists from the men's 100 metre final ranged from 4.70 to 5.04 m at the 70 metre mark (Ae, Ito, and Suzuki, 1992). This may indicate that stride length is a somewhat individual characteristic, even in elite sprinters. For women, values from 3.62 to 4.34 m have been reported for internationally ranked sprinters (Levtshenko, 1990). Similar values of 3.74 to 4.04 m were found in German and American female sprinters in an international dual meet (Baumann, 1985). These values may be used as a guide for "normal" stride lengths of male and female sprinters, but there was no indication in these reports of the heights of the athletes and therefore it is incorrect to state that these values are ideal for all sprinters. Mero, Luhtanen and Komi (1986) reported step lengths of 2.16 m for male sprinters (mean height = 1.80m), and 1.91m for female sprinters (mean height = 1.67m). The value for males is noticeably lower than those of Nummela, Vuorimaa, and Rusko (1992), and Hoskinsson and Korchemny (1991), but the sprinters in this study were not elite level (mean 100m time = 10.84 sec), so their step length would expectedly be less. Hoffman (1971), in performing a regression analysis on sprinter's height and step length, found the maximum step length of male sprinters with personal best 100 metre times of 10.7 seconds or less is equal to 1.265 times the athlete's overall standing height. This is similar to Chengzhi (1991) who found that the average step length of the eight finalists for the men's 100 metres at the 1988 Olympics was 1.24 times the average height of the athletes.

As the speed of running increases, the flight time increases and the contact time decreases (Adrian & Cooper, 1989). Hay (1993) reported the time in the contact phase may be as low as 40% to 45% of the total step time. In a kinematic study of the men's 200 metres at the 1984 Olympics, Mann and Herman (1985) found the gold and silver medallists were in contact for 43.4% and 45.8% of step time, respectively. They also found the eighth place finisher in the race was in contact for 52% of step time, which suggests that the time period of ground contact is important as it is indicative of velocity the sprinter can generate (Mann, 1986). Hay and Reid (1988) reported that the time of the contact phase in elite sprinters is approximately 0.07 to 0.09 seconds when running at maximum speed. This is similar to the contact time of 0.10 seconds found by Burt (1994), and the 0.09 seconds reported by Mero, Luhtanen and Komi (1986) in studies of elite sprinters. In a study comparing elite level and collegiate level sprinters, Mann and Herman (1985) found significant differences in the stride frequency (elite higher) and contact time (elite lower), with no significant differences found in stride length or flight time. These results suggest which factors may be more important in sprinting success, that a greater stride frequency and a shorter support time can increase performance, and that improving stride length and flight time may not result in faster sprinting speeds. According to Mero, Luhtanen, and Komi (1986), top female sprinters achieve a stride frequency of 2.28 strides per second when running at maximum speed. This is similar to the 2.24 strides per second found by Hoffman (1971), which is notable considering the difference in date between publications. For males, Hoskisson and Korchemny (1991) reported stride frequencies up to 2.59 strides per second in elite junior sprinters. Stride frequencies of the eight finalists of the 100 metres at the 1991 World Championships in Athletics ranged from 2.27 to 2.48 strides per second (Ae, Ito, and Suzuki, 1992) at the 70 metre mark. Mann (1985) reported "good" male sprinters achieve a stride rate of 2.4 strides per second, while "average" male sprinters achieve 2.25 strides per second and "poor" male sprinters achieve 2.1 strides per second. It may be expected that elite level male sprinters will have a greater stride frequency than females, as they have a faster running velocity, and possess greater strength and power.

2.2.4 Lower Body Kinematics

The keys to successful sprinting lie in the kinematics of the lower body, as it is the movements of the legs throughout the sprint stride which influence the ground reaction forces generated during ground contact to produce high running speeds. According to Mann (1986), improving sprint performance is seen in the leg action immediately prior to and during ground contact, as it was found that elite sprinters minimize hip range of motion during ground contact and produce greater hip extension angular velocity during support. Ground contact time is decreased by generating fast hip extension angular velocity prior to ground contact, touching down with a smaller horizontal distance from the contact foot to the body centre of mass, maintaining the hip extension angular velocity during ground contact, and by leaving the ground as quickly as possible. Early researchers (Bunn, 1978, Dillman, 1975) have stated that one of the most common errors in sprinters is incomplete knee extension at take-off and after take-off. This is in contrast to Mann (1985), who found that in elite sprinters there was a lack of full knee extension at toe-off, which helps to minimize ground contact time (Figure 2-4). Kinematic studies confirm this; in which maximum knee extension angles of 155.7° (Hoskisson & Korchemny, 1991) and 165° (Tupa, Dzhalilov, & Shuvalov, 1991) as measured between the thigh and shank segments have been noted. As stated by Mann (1986, p.3001) "in the possible tradeoff of greater leg extension to increase speed versus abbreviated leg extension to decrease ground contact time, it appears that the latter produces better results."

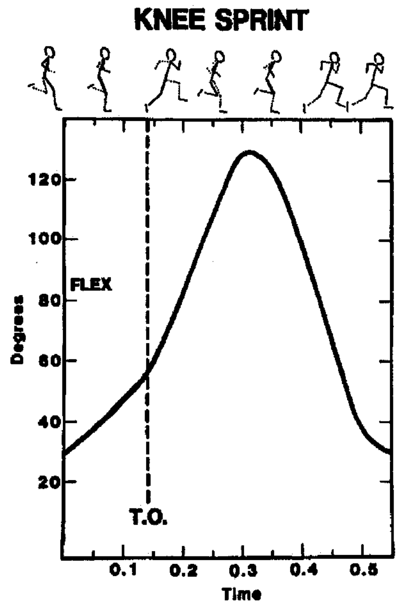


Figure 2-4. Range of motion at the knee during sprinting. Knee angle in this figure is measured as 180° minus the angle between the thigh and shank segments (Mann et al., 1986, p. 505).

After the contact foot leaves the ground, the entire recovery leg must rotate forward, accelerating to catch up and pass the body, and then rotate back in order to push against the ground again. The faster the horizontal velocity of the sprinter, the faster the leg must recover and push. An increased recovery speed is accomplished by more forceful hip flexion torque and by increased flexion at the knee and dorsiflexion at the ankle. These movements act to decrease the radius of gyration about the hip, thus decreasing the moment of inertia of the leg. According to Dare (1984), the amount of knee flexion of the recovery leg is to some extent an individual characteristic of a sprinter, depending on individual morphology. Some runners minimize the knee flexion angle in which the foot comes into contact with the buttocks, while in others it is less pronounced and the foot simply follows the action of the knee and is swung forward. Ideally, the knee angle should be minimized for each sprinter, as it will decrease the moment of inertia (Tupa, Dzhalilov, & Shuvalov, 1991). Hoskisson and Korchemny (1991) found the minimum knee flexion angle to be 32.5° as measured between the thigh and shank segments. This is somewhat smaller than the 38.7° found by Tupa, Dzhalilov, and Shuvalov (1991) which may indicate that the amount of knee flexion is indeed an individual characteristic of sprinters and is a function of both sprint mechanics and individual morphology.

In a study of internationally ranked Canadian and American sprinters, Lemaire and Robertson (1990) found peak knee flexion angular velocities of 1030°/s. Chengzhi and Zongcheng (1987) found larger knee flexion angular velocities of approximately 1400°/s in sprinters with personal best 100 metre times of 10.0 to 10.1 seconds. These knee angular velocity values occur as a result of hip flexion during the forward swing of the recovery leg. This knee flexion occurs passively, as EMG studies (Wiemann & Tidow, 1995; Mann, et al., 1986) have found that there is no activity of the hamstring muscles during this knee flexion.

As the foot accelerates ahead of the body, hip flexion occurs and the thigh on the contralateral side is driven forwards and upwards. It is this forceful hip flexion which increases the forces applied to the ground, thus increasing the ground reaction forces which act to propel the sprinter forward. The greater the hip flexion torque, the greater the ground reaction force. In an article by Dare (1984), hip flexion was incorrectly described as being a result of the ground reaction forces imparted to the leg in contact with the ground The author stated that the greater the ground reaction force, the greater the running speed, and therefore more hip flexion. Hip flexion does not occur as a result of the ground reaction force; it is the flexion of the hip that contributes to ground reaction force production.

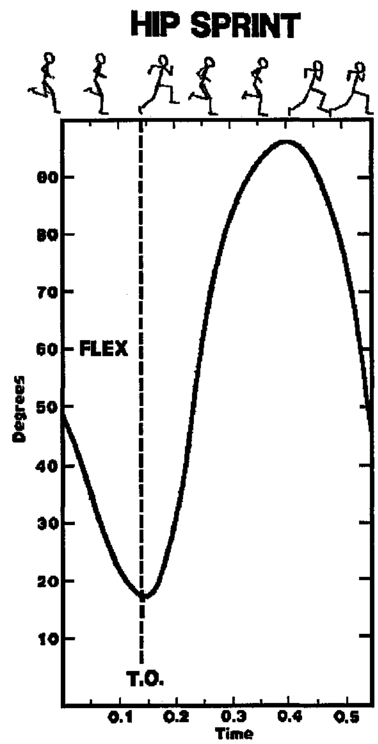


Figure 2-5. Range of motion at the hip during sprinting. Hip angle in this figure is measured as 180^{O} minus the angle between the trunk and the thigh. (Mann et al., 1986, p. 504).

Mann et al. (1986) found the minimum angle of hip flexion between the trunk and the thigh to be 100° (Figure 2-5), with Hoskisson and Korchemny (1991) reporting a similar value of approximately 101.2° in elite junior sprinters. Maximizing the angle of hip flexion is a necessary component of fast sprinting, as it helps to ensure the production of hip extension angular velocity (Mann & Herman, 1985).

Chengzhi and Zongcheng (1987) found a peak hip flexion angular velocity value of approximately 900°/s. This is similar to the 969°/s reported by Lemaire and Robertson (1990), but is considerably larger than the hip flexion angular velocity value of almost 600°/s reported by Mann (1985). The large differences between these angular velocities are interesting, as the subjects in each of these studies were elite level sprinters, and demonstrated similar minimum knee flexion angles of approximately 33° (Chengzhi & Zongcheng, 1987) and 31° (Mann, 1985).

The horizontal distance between the foot at the moment of ground contact and the centre of mass is a strong predictor of sprinting performance (Alexander, 1989). Ground contact should occur as closely beneath the centre of mass as possible (Deshon & Nelson, 1968), with distances of 6-8 centimetres from foot contact to centre of mass reported (Mann, 1985). The closer the ground contact occurs beneath the centre of mass the smaller the horizontal braking force will be which slows down the sprinter, however, even when the foot is placed directly beneath the centre of mass it will still not prevent unwanted braking (Payne, Slater, & Telford, 1968).

Hip extension angular velocity should be maximized in sprinting prior to ground contact in an attempt to minimize the linear velocity of the foot at the instant of ground contact, as the horizontal velocity of the foot determines if there is a braking effect when the foot contacts the ground (Hay, 1993). For example, in Figure 2-6(a), the horizontal velocity of the centre of mass of the sprinter is 10 m/s as defined in the inertial frame of reference, while at the instant of ground contact the horizontal velocity of the foot is 2 m/s. This means the foot is moving forwards relative to the ground at contact, resulting in a braking force at ground contact. In Figure 2-6(b), the horizontal velocity of centre of mass is 10 m/s, whereas the horizontal velocity of the foot at the moment of ground contact is 2 m/s in the opposite direction. This means the foot is moving backwards relative to the ground at the moment of contact which would result in a propulsive force at ground contact.

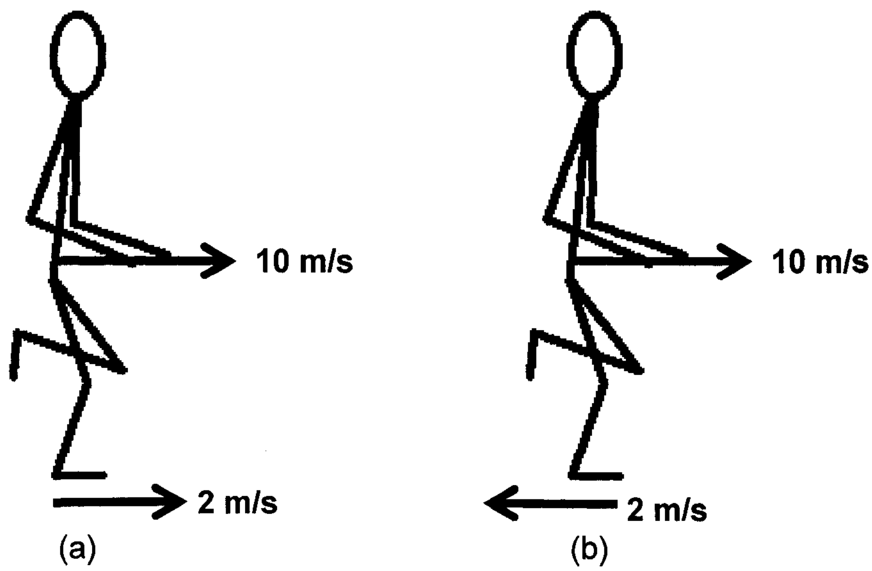


Figure 2-6. Horizontal velocity of the foot at the instant of contact relative to the global coordinate system. a) The foot is moving forwards with a velocity of 2 m/s, resulting in a braking force at contact. b) The foot is moving backwards with a velocity of 2 m/s, resulting in a propulsive force at contact.

According to Mann (1985), no sprinter has been able to recover the foot so that it is moving backwards relative to the ground at the moment of contact. Mann and Herman (1985) found foot horizontal velocities at the moment of ground contact of 2.28, 4.09, and 2.82 m/s in the first, second, and eighth place finishers in the men's 200 metres at the 1984 Olympics. Mann (1985) reported that "good" male 100 metre sprinters attained horizontal foot velocities of approximately 1.7 m/s at ground contact, while "average" sprinters achieved approximately 2.6 m/s and "poor" sprinters achieved approximately 3.5 m/s. Better sprinters are able to minimize the braking force on ground contact by decreasing the horizontal velocity of the foot relative to the ground at the instant of ground contact.

Peak hip extension angular velocities of 912°/s have been reported by Lemaire and Robertson (1990) during the support phase, which are much larger than the values of approximately 600°/s (Chengzhi & Zongcheng, 1987) and 500°/s (Mann, 1985) which have been reported during support in other studies.

Peak knee extension angular velocities during recovery of 1200°/s (Lemaire & Robertson, 1990) and approximately 1300°/s (Chengzhi & Zongcheng, 1987) have been reported. Knee extension occurs passively, as Wiemann and Tidow (1995) and Mann et al. (1986) found that there is minimal EMG activity in the quadriceps during this portion of recovery. These values were found late in the recovery phase, and occurred in association with the deceleration of hip flexion followed by rapid hip extension (Figure 2-7). This passive knee extension is due to the summation of speed principle, which states that body segments move in sequence, starting with the more proximal segments and ending with the more distal segments, and the motion of each segment should begin when the preceding segment has reached its maximum speed. The summation effect is such that the more distal the segment, the faster it will eventually move (Dyson, 1986).

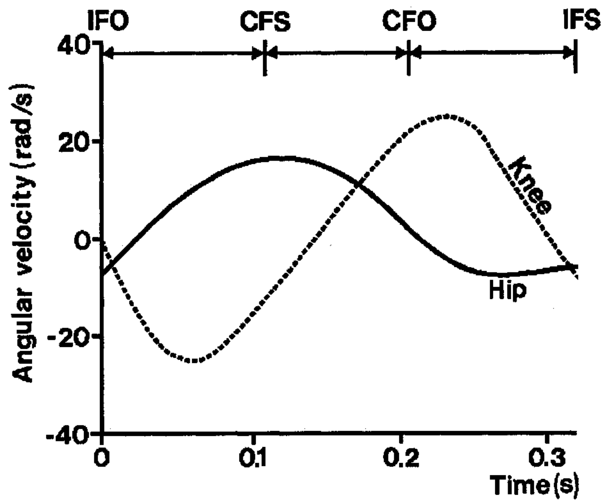


Figure 2-7. Angular velocity curves of the hip and knee for the swing leg during recovery. IFO = ipsilateral foot take-off; CFS = contralateral foot strike; CFO = contralateral foot take-off; IFS = ipsilateral foot strike (Chengzhi & Zongcheng, 1987, p. 826).

At the ankle, the foot is primarily plantarflexed throughout the sprinting stride (Figure 2-8). It is only during the support phase when the body passes over the foot in contact with the ground that dorsiflexion occurs, with a value of approximately 10° relative to a neutral ankle position (0°). Maximum plantarflexion occurs at the toe-off, where values of approximately 24° relative to a neutral ankle position (0°) have been reported (Mann et al., 1986) in male sprinters, although the ability of these athletes was not indicated. Hoskisson and Korchemny (1991) reported maximum plantarflexion values in elite junior sprinters of 14.2°. A decreased range of plantarflexion may be more desirable in sprinting as limiting the amount of plantarflexion may help decrease the contact time. After toe-off, dorsiflexion occurs at the ankle, but the foot is only brought approximately to a neutral position. This action helps to decrease the moment of inertia or the resistance to angular motion of the recovery leg, allowing it to swing forward faster for a given torque.

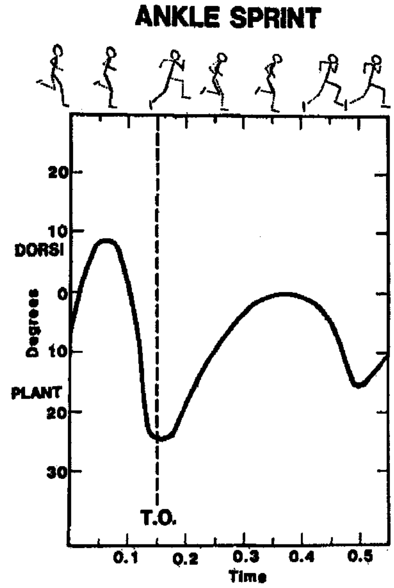


Figure 2-8. Range of motion at the ankle during sprinting (Mann et al., 1986, p. 507). Dorsiflexion and plantarflexion angles are measured relative to a neutral ankle position (0°).

2.3 Kinetics of Sprinting

Kinetic studies are those in which the forces causing movement or changes in movement are investigated. The forces involved in movements provide an understanding of motion on a more complex level. Kinetic studies in sprinting are relatively new, starting in the early 1980's (Mann & Sprague, 1983). According to Mann (1985), kinetic analyses are an excellent measure of sprinting quality, but are difficult to perform accurately.

One important point regarding kinetic analyses is that the results indicate the muscle groups that are dominant in the activity. Conclusions can be reached regarding the general muscle action, but due to the nature of the analysis methods specific muscle group activity cannot be determined (Mann, 1981). The moment calculation assumes that during any motion, the antagonist muscle or muscle groups are inactive and not producing any force of their own (Gagnon, Robertson, & Norman, 1987). Therefore, if the antagonist muscle groups were active during the moment calculations, the force produced would be greater than the values obtained.

2.3.1 Kinetics of the Hip

In sprinting, the hip flexors contract eccentrically to stop the posterior rotation of the thigh just prior to toe-off, and begin anterior thigh rotation by concentric contraction of the hip flexors (Dillman, 1971; Mann & Sprague, 1983). Once this has been accomplished, there is a brief period when the contribution of the hip flexor muscles is minimized and there is a small angular acceleration of the thigh. This occurs as a result of knee flexion during the swing phase decreasing the moment of inertia of the leg, allowing it to swing through more easily and rapidly without muscular assistance. Following this brief period of decreased muscle activity, the hip extensors are recruited to stop the anterior thigh rotation by contracting eccentrically, producing an extensor moment. This extensor moment continues through to ground contact, where in the resistive phase a high extensor moment is generated. This muscular activity, which has been related to the incidence of hamstring injury (Iiboshi, Ae, Suenaga, & Miyashita, 1987), is necessary to minimize the horizontal braking force produced during this portion of ground contact. As the contact phase progresses, there is a flexor moment as the muscle dominance shifts to the hip flexors which decrease the backwards rotation of the leg, and initiate forward swing (Figure 2-9).

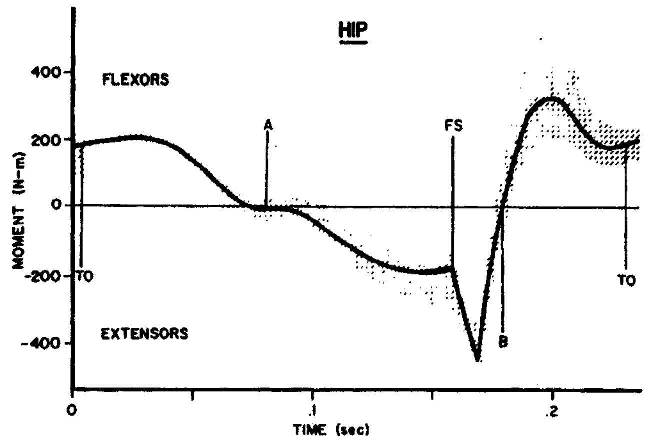


Figure 2-9. Average moment pattern generated at the hip during one complete sprint stride. (Mann, 1981, p.327). TO = take-off; A = ankle cross; FS = foot strike; B = centre of mass above base of support

Peak hip moment values for male world class sprinters were found to be approximately 410 Nm in the hip extensors and 305 Nm in the hip flexors during sprinting (Mann & Sprague, 1983). Similarly, hip extensor moments of 380 Nm and hip flexor moments of 290 Nm were found by liboshi et al. (1987) in uninjured male sprinters in a study comparing flexion and extension moments in injured and uninjured sprinters. These values are larger than those found by Alexander (1990) in a study of national level sprinters, where peak hip moment values were 364 Nm in hip extension and 241 Nm in hip flexion. The differences in these values can be attributed to the fact that the sprinters in the study by Mann and Sprague (1983) were at a higher level of sprinting, and the kinetic data were collected while the athletes were sprinting at maximum velocity, where the angular velocity of the hip in flexion and extension was approximately 600°/s and 400°/s, respectively. This means greater muscle moments were required to decelerate the limb in one direction, and start it moving in the other direction. In comparison, the subjects in the study by Alexander (1990) were at a lower level of sprinting and results were collected using an isokinetic dynamometer, and therefore were not a true representation of the muscular forces involved in sprinting as the maximum angular velocity tested at the hip in flexion and extension was fixed at 180°/s. It must be noted that the peak moment values reported by Mann and Sprague (1983) and Iboshi et al. (1987) were seen during the contact phase. Two studies have reported hip kinetics specifically during the recovery phase. Vardaxis and Hoshizaki (1989) examined power patterns of the lower extremity during the recovery phase of intermediate and advanced level sprinters. Hip flexor moments of 300 Nm and extensor moments of 350 Nm were found in advanced sprinters (Figure 2-10). Dillman (1971) performed a kinetic analysis of the recovery leg during sprinting. Hip flexor moments of greater than 300 Nm and hip extensor moments of approximately 300 Nm were reported.

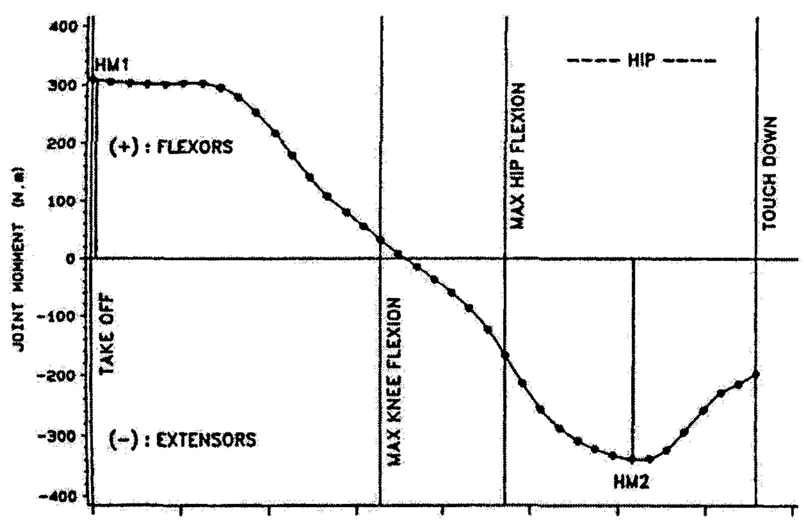


Figure 2-10. Hip moment during recovery phase in advanced sprinter (Vardaxis and Hoshizaki, 1989, p. 342).

2.3.2 Kinetics of the Knee

At the beginning of the flight phase, the knee muscle moment is dominated by the eccentric action of the knee extensors, which are working to halt knee flexion (Dillman, 1971). This means that knee flexion during recovery is limited by the knee extensors, not initiated by the knee flexors (Mann, 1981). Once knee flexion has been stopped, the knee extensor moment then functions to extend the knee. As knee extension progresses, there is a brief period of minimal muscle activity when the lower leg swings forward due to the angular momentum generated when the thigh is accelerated forward. As the lower leg approaches maximum knee extension, the knee flexors begin to contract eccentrically to slow the lower limb. This is quickly followed by a concentric contraction of the knee flexors to accelerate the lower leg backwards into ground contact. At footstrike, the flexor activity continues briefly to decrease the horizontal braking force which results from ground contact in front of the centre of mass (Hay, 1993). As the sprinter approaches the support phase of ground contact, the knee extensors begin to contract eccentrically, stopping the negative vertical velocity of the body as the centre of mass reaches the lowest point of its parabolic arc. After the centre of mass passes the foot and the sprinter enters the propulsive phase, the knee flexors begin to contract concentrically to produce positive vertical and horizontal velocity. As toe-off is approached, the knee extensors decrease in muscular activity to protect the rapidly extending joint from harmful hyperextension (Iiboshi et al., 1987) (Figure 2-11).

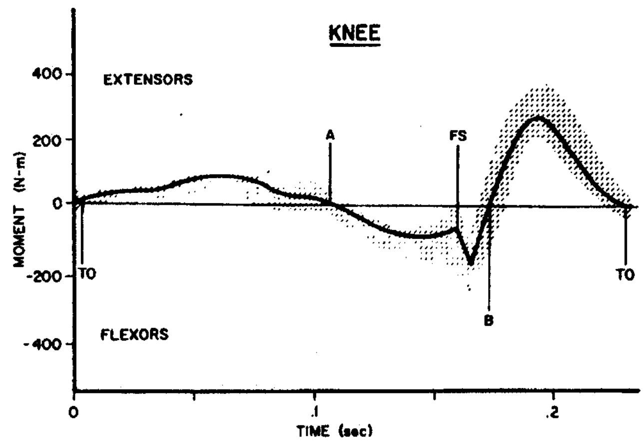


Figure 2-11. Average moment pattern generated about the knee during one complete sprint stride. (Mann, 1981, p.326). TO = take-off; A = ankle cross; FS = foot strike; B = centre of mass above base of support.

Mann and Sprague (1983) reported peak knee moment values for male elite level sprinters of approximately 300 Nm in knee extension and 195 Nm in knee flexion during sprinting. These values were comparable to those found by Iiboshi et al. (1987) of 290 Nm in knee extension and 160 Nm in knee flexion. They were also similar to those found by Alexander (1990) when using an isokinetic dynamometer for data collection, where peak moment values of 276 Nm in knee extension and 176 Nm in knee flexion were reported. As was the case at the hip, the peak values of Mann and Sprague (1983) and Iboshi et al. (1987) were found during the contact phase. During the recovery phase, Vardaxis and Hoshizaki (1989) showed knee extensor moments of 100 Nm and knee flexor moments of 200 Nm in advanced sprinters. Dillman (1971) reported similar values.

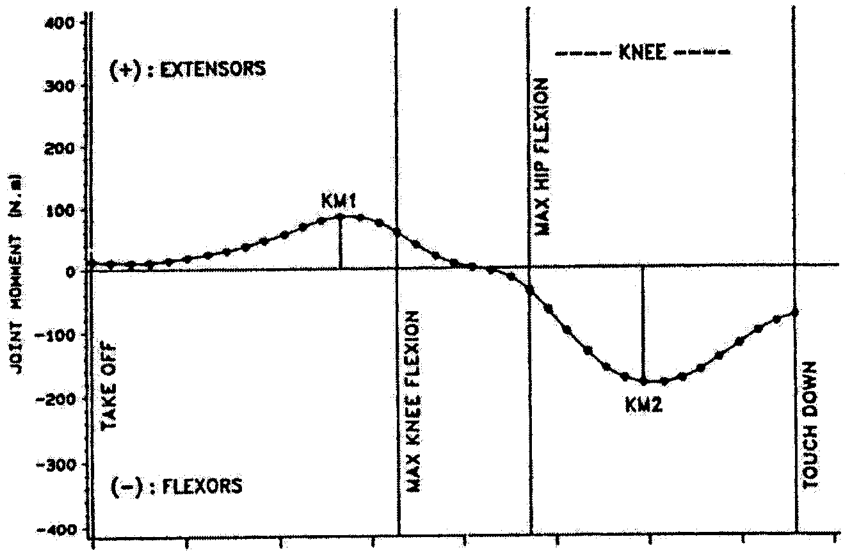


Figure 2-12. Knee moment during recovery phase in advanced sprinter. (Vardaxis and Hoshizaki, 1989, p. 343).

2.3.3 Kinetics of the Ankle

The muscle moment at the ankle during the flight phase is minimal (Dillman, 1971), which indicates that the resultant joint moment associated with muscular activity during this phase is balanced (Mann, 1981). During the resistive and supportive phases of ground contact, the plantarflexors contract eccentrically to halt the negative vertical velocity of the body as the centre of mass of the sprinter reaches the lowest point on its parabolic arc. Once the sprinter enters the propulsive phase, the plantarflexors contract concentrically to produce positive vertical and horizontal velocity to project the body into the flight phase. The large resultant joint moment at the ankle seen during the resistive and supportive phases demonstrates the strength capabilities of a muscle group when working eccentrically. The rapid decrease in moment magnitude once the concentric contraction is initiated in the propulsive phase, however, suggests that the belief that the importance of the plantarflexors during the latter stages of ground contact may be overemphasized (Mann, 1981) (Figure 2-13).

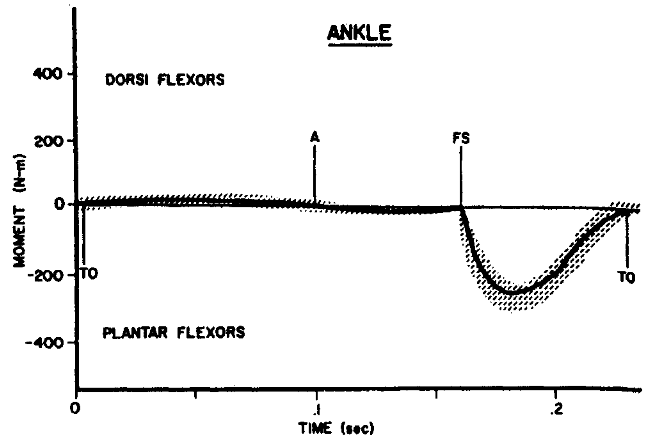


Figure 2-13. Average moment pattern generated about the ankle during one complete sprint stride. (Mann, 1981, p. 326). TO = take-off; A = ankle cross; FS = foot strike.

At the ankle, Mann and Sprague (1983) found peak moment values of approximately 20 Nm in dorsiflexion and 230 Nm in plantarflexion in a study of elite sprinters. Comparatively, Alexander (1990) found peak moment values of 47 Nm in dorsiflexion and 127 Nm in plantarflexion. The small moment values for the dorsiflexors in both studies is representative of their relatively minor contribution during sprinting, as they function to dorsiflex the ankle during the recovery phase. The difference between the peak plantarflexion values reported is due to the fact that Alexander (1990) measured the joint moments using an isokinectic dynamometer, whereas Mann and Sprague (1983) calculated the resultant joint moments during ground contact of sprinting using a force platform. The dynamic lower extremity motion during sprinting resulted in a larger plantarflexor moment at the ankle as compared to the controlled (150°/sec) measurements using an isokinetic dynamometer.

Success in producing high velocities during sprinting is dependent on the ability to produce large muscle moments in the lower extremity throughout the sprint race. Mann and Sprague (1983) suggested that the muscular activity at the hip makes the greatest contribution to success in sprinting. During ground contact, the hip extensor moment required to continue hip extension is major contributor to performance. After toe-off, the hip flexor activity needed to recover the leg forwards and the hip extensor activity required to slow leg recovery and extend the limb toward ground contact are also important to success. At the knee, it appears as though the flexor moment prior to ground contact, the knee flexor activity just after contact, and the extensor moment during ground contact also factors related to successful sprinting. In addition, the level of activity of the plantarflexors is important during the resistive and support phases of ground contact. Mann and Sprague (1983) did not indicate why the knee or ankle were significant contributors to sprinting success.

Chapman and Caldwell (1983) investigated the kinetic limitations of maximal speed sprinting. Studying one female elite level sprinter on a treadmill over a range of velocities, they found that there was an increase in the peak total leg energy at submaximal speeds but little change between the penultimate and maximal speed. Consequently, at high speeds leg recovery was completed successfully by delaying the reduction of total leg energy prior to ground contact. It appeared as though the extent of this delay was the limiting factor between the penultimate and the maximum speed, and it was the inability to increase the peak eccentric knee moment which limited this delay. This made it necessary to spend more time in the flight phase and resulted in a modification of the relationship between stride frequency and stride length at high speeds, with the stride frequency decreasing as the stride length continued to increase. The authors recognized that the results of this study were only applicable to one athlete, but they believed that the changes observed represented an approach to the understanding why speed is limited (Chapman & Caldwell, 1983).

2.4 Computer Modeling and Simulation

Computer modeling has been described as "the setting up of mathematical equations to describe the system of interest, the gathering of appropriate input data, and the incorporation of these equations and data into a computer program" (Vaughan, 1984, p. 373). Similarly, computer simulation can be defined as "the use of a validated computer model to carry out 'experiments,' under carefully controlled conditions, on the real-world system that has been modeled" (Vaughn, 1984, p. 373). It is feasible for computer modeling to be performed without the subsequent computer simulation taking place, however, it is not possible for the simulation to occur without the development of the computer model (Vaughan, 1984).

There are a number of advantages to using computer simulation in sport biomechanics research. The first is safety, as there are no hazardous experiments for the athlete. The second advantage is time, as many different simulations can be performed quickly. Thirdly, there is minimal expense with computer simulation, as there is no need to build different physical models. With the development of powerful personal computers and specialized software, computer simulation has become easier to perform. Finally, computer simulation allows for the prediction of an optimal performance for a given skill (Vaughan, 1984).

There are a number of disadvantages to computer simulation as well. Validation of the model is difficult, and incorrect models will prejudice the results of the experiment. In addition, advanced mathematics and computer programming skills are often required of the user. Also, the results of the simulation are often difficult to translate into practical terms.

According to Miller (1979), biomechanical models may be divided into two groups, physical models and mathematical models. Most biomechanical models of sports skills and locomotion are basically mathematical models of a rigid body-inertial parameter type, and may be either planar or spatial. Planar are commonly adapted for motions which are reasonably symmetrical in the sagittal plane because of their greater simplicity.

Most dynamic rigid body models are based upon either Newtonian or Lagrangian equations of motion. For the Newtonian approach, there are two vector equations:

$$\Sigma F = m\, a_o$$

and

$$\Sigma M = I_o\, \alpha$$

That is, the sum of the forces equals the mass times acceleration and the sum of the moments of force equals the moment of inertia times the angular acceleration. Subscript o indicates the centre of mass. The second major mathematical approach to dynamic rigid body models incorporates Lagrangian mechanics. The Lagrangian principle states that of all possible paths accessible to a dynamical system, the one chosen by nature is that which minimizes the time integral of the scalar quantity

$$L = T - U$$

in which T and U are the kinetic and potential energies of the system and L is defined as the Lagrangian. The equations of motion deal with energy as the dependent variable and generalized coordinates as independent variables. The same numbers of differential equations are derived as there are degrees of freedom within the system.

Simulation models vary in their level of sophistication and complexity, ranging from simple models (e.g. Farley & Gonzalez, 1996) to very complex models (e.g. Hatze, 1981a). Farley and Gonzalez used a simple spring-mass model which was comprised of a single linear leg spring and a mass equivalent to the system's mass to illustrate stride frequency and leg stiffness in human running. In comparison, Hatze (1981a) used a very sophisticated 17 segment model to simulate the takeoff during the long jump. The model incorporated a very complex muscle model to represent each of the 46 muscle groups, and required the activation of each muscle group in addition to the segment configurations and orientations. Inertia parameters were determined for the 17 segments using 242 anthropometric measurements (Hatze, 1980), with muscle parameters estimated from torque and EMG measurements during maximum isotonic and isometric contractions (Hatze, 1981b).

Alexander (1992) believed that the human body was far too complex to be replicated in detail by any model. All human body models devised by biomechanists are simply representations of the real system. Alexander stated that it is not always necessary for simulation models to include as much complexity as possible. Simple models may not be appropriate in certain circumstances, but it is generally advisable to keep the model as simple as is required by the research question. Hubbard (1993) stated that one should always begin with a model which has a simplified design yet still captures the essence of the question being investigated. From this basic model, additional complexity should be added only when it is necessary, such as when it is clear that the model cannot represent the complexity of the behaviour. One must be aware, however, that an overly simplified model will be restricted in its applicability, may omit key features, and may have poor accuracy (Yeadon & Challis, 1994). Models are most effective when they are based on and demonstrate fundamental physical principles, rather than simply being a mathematical representation involving a number of randomly chosen coefficients which have no theoretical basis or justification. Physically based models will inevitably yield more understanding than those which are not based on physical concepts (Hubbard, 1993).

Various software programs have been developed to facilitate the development of simulation models, including Working Model 2D, Matlab, Autolev, and AnyBody. Working Model 2D has many features which allow for ease in developing and running simulation models. Geometric solids may be used to represent the various segments of the model, and may be specified for length, weight, and inertial parameters. Joints may be represented by frictionless pins, with torque generators located at the joints to produce movement, with the input values read via Dynamic Data Exchange (DDE) from programs such as Microsoft Excel or Matlab.

2.4.1 Modeling, Simulation, and Optimization in Human Locomotion

The gait cycle is one of the most common and important activities of daily life, as well as playing a key role in sporting movements. The computer modeling of human locomotion has received a great deal of attention from biomechanics researchers (Vaughn, 1984); accordingly various simulation models of gait and human locomotion have been published.

When modeling gait and locomotion, several investigators have employed optimal control theory, utilizing a criterion of minimum energy in their studies (e.g., Beckett & Chang, 1968; Chow & Jacobson, 1972; Nubar & Contini, 1961; Seirig & Arvikar, 1973). This approach, however, is not appropriate for movements which require maximum effort (Hatze, 1976), as is the case with sprinting. In such circumstances, optimization techniques have been used to investigate the optimal temporal sequencing of limb movements (Pandy & Zajac, 1991; Bobbert & van Ingen Schenau, 1988, Hatze, 1976) or optimal muscle coordination strategies (Pandy & Zajac, 1991; Spagele, Kistner, & Gollhofer, 1999). Opinions differ when it comes to the applicability of optimal control with human movement. Mena, Mansour, and Simon (1981) believed that optimal control studies may be mathematically elegant, but their focus has been on the optimal nature of human gait rather than examining the observable and alterable physical variables which produce the overall movement pattern. In contrast, Pandy, Zajac, Sim, & Levine (1990) stated that optimal control is "currently the most sophisticated methodology available for solving human movement synthesis problems" (p. 1186). These varying opinions may be partly due to the fact that there is a lack of consistency in the sophistication of the muscle models used. Audu and Davy (1985) showed that the complexity of the model is very influential in determining the predicted kinematic and kinetic outcomes of the problem. When applied to sports movements, more appropriate muscle models still need to be established (Yeadon & Challis, 1994) which accurately represent the athlete's muscle model parameters.

The spring-mass model, which consists of a massless spring attached to a point mass, has been used to characterize human running. Its application is based on the theory that muscles, tendons, and ligaments all behave like springs, and during running these complex musculoskeletal springs function like a single linear spring (Farley & Gonzalez, 1996). Despite the fact that it is a simplification of the musculoskeletal system, this type of model describes running remarkably well (Blickhan, 1989). The spring-mass model has been used to determine the relative importance of altering the stiffness and the angular displacement of the leg as running speed changes (Farley & Gonzalez, 1996), as well as the relationship between stride characteristics and ground reaction force production during running (Derrick, Caldwell, & Hamill, 2000). Whittlesey and Hamill (2004) pointed out that interpretation of the data from the spring-mass model is not straightforward because it is somewhat abstract. Because of the nature of the model, the mass cannot be distributed accurately. In addition, humans actually use multiple joints when absorbing impacts during running so it is difficult to draw a direct comparison to a model which incorporates fewer joints. The lower extremity clearly does not function as a spring, but important information can be obtained from such a simple model.

Inverse dynamics models are those in which the kinematics, obtained experimentally, serve as system inputs and solutions are found for the moments and forces applied to the system (Seigler, Seliktar, & Hyman, 1982). This method had been used by Thornton-Trump and Daher (1975) to predict the ground reaction forces during human locomotion, and by Seirig and Arvikar (1975) to estimate joint reaction forces and muscular forces during gait. When applied to optimal control problems, Happee (1994) believed that inverse dynamics has an advantage over numerical methods in that the computational complexity is greatly reduced because the solution for each position can be computed independently of the solution for other samples. This method has been combined with optimization techniques to construct a kinematically and kinetically correct model from which ground reaction forces and joint moments have been estimated (Koopman, Grootenboer, & de Jongh, 1995).

A direct dynamics model requires forces and moments to serve as the system inputs and the solutions are found for the system kinematics (Seigler, Seliktar, & Hyman, 1982). Adjustments can then be systematically made to the input parameters and the resulting changes to the kinematics and kinetics of the system can be quantified. Mena, Mansour, and Simon (1981) used this approach to model the leg during the swing phase of gait to evaluate the affects of changing various parameters related to the gait cycle, including adjusting the initial conditions of swing and limiting joint range of motion. Onyshko and Winter (1980) developed a seven segment model in which the limb angles and velocities were the initial conditions, and the joint moments were the system inputs. Kinematic and kinetic data for the simulation were obtained from gait measurements. The model progressed through the normal walking cycle, and atypical gait patterns were demonstrated with minor perturbations to the data. A direct dynamics simulation using a musculo-skeletal model of the lower extremity was performed to determine the effect of muscle activation, body positions and initial velocities, and surface properties on the impact phase of running (Gerritsen, van den Bogert, & Nigg, 1995).

Direct dynamics models, as is the case with others, can vary greatly in their complexity. The model by Mena et al. (1981) used torque values as their input, whereas Gerritsen et al. (1995) incorporated sophisticated muscle models which included contributions from individual muscles and various physiological properties of muscle. The difficulty comes, however, in ensuring the accuracy of the muscle model. If muscle-specific parameters are to be used they must be determined experimentally, which is not feasible when studying elite athletes and data is collected during competition. When it comes to analyzing sports movements, it may be more appropriate to use a torque-based model (Yeadon &

Challis, 1994), as the input data can be more easily and accurately determined through experimental methods.

2.4.2 Modeling, Simulation, and Optimization in Sprinting

A number of studies have been published related to modeling sprinting, however most deal with performance in a race as compared to modeling the mechanics of the sprint stride. A series of theoretical experiments have been completed simulating the velocity curves for running events of different distances, using the sprinter as a single system. Keller (1973, 1974) derived velocity curves for sprinting and running races from 50 m to 10000 m. In developing the model, the author made the assumption that the propulsive force during sprinting was constant because there is a maximum effort for the entire race. In addition, it was assumed that resistive forces were proportional to velocity. However, it was not clear whether these forces were internal, external, or both. This model predicted that an athlete should be able to sprint at top speed for close to 300 m, and that the optimal race strategy for longer races would require the runner to slow down during the last few meters. Accurate predictions were seen from this model, with less than 3% error.

Vaughan (1983a, 1983b) and Vaughan and Matravers (1977) argued that there is a decrease in the driving force during sprinting with an increase in velocity, which does not remain constant during a race. In addition, these researchers proposed that the air resistive force should vary with the square of the velocity, as is consistent with the principles of fluid mechanics. The strengths of their model were seen in the fact that it could predict a sprinter's time for a particular distance, it could be used to pinpoint where the sprinter's strengths and weaknesses may be found, and that it could be applied to sprinters of both genders. However, one of the model's weaknesses was that it was limited to sprinting and did not attempt to model any of the physiological parameters that are influential on performance. In addition, the resulting velocity and acceleration curves for the 100 metres did not show any deceleration phase towards the end of the race. In reality, the sprinter is not able to maintain maximum velocity for the entire duration of the race; therefore the model begins to break down at some point.

Keller's theory of sprint running was further developed by Senator (1982), with the ultimate goal being to achieve a better agreement between Keller's estimate of maximum steady-state rate of developing energy in a runner's muscles and measured oxygen uptake. Senator assumed the resisting force to be proportional to any positive power n of the velocity (Keller used $n = 1$, Vaughan used $n = 2$) and allowed the driving force to vary during the race. However, his strategy for varying the driving force was not consistent with experimental findings (Baumann, 1976), since he argued that the sprinter would have to exert greater forces as sprinting speed increased. Perhaps the most significant conclusion drawn by Senator was that accurate displacement-time data would have to be made in the critical acceleration region (0 to 20 m) to determine a definitive value for n.

Phillips, Roberts, and Huang (1983) used a two segment model to quantify the non-muscular reactions between two adjacent segments undergoing free-segment motion, as is seen in the lower extremity during the recovery phase of running or sprinting. A Newtonian approach was used to simulate the trajectories of the two segments, with the muscular moment acting only about the proximal end of the proximal link and the linear acceleration as the input parameters. The intersegmental forces and moments were considered to be zero, and movements of the two segments were not constrained in any way. The simulated motion of the thigh and shank segments during the swing phase of running was comparable to the data obtained from the one participant. Three initial conditions were considered: (1) the thigh segment was starting to rotate forward; (2) the thigh segment was rapidly rotating forward, but maximum angular velocity was not yet reached; and (3) the thigh segment reached maximum negative acceleration. The results showed that as the thigh forward rotational speed increased there was a corresponding increase in knee flexion angular velocity. Also, when the speed of the thigh's forward rotation decreased later in the swing phase, knee extension occurred passively without knee extensor musculature involvement.

Mathematical modeling and optimization was used by Wood, Marshall, and Jennings (1987) to study the role of the hamstring muscle group during sprint running. The objective was to identify the principles which determine the movements of the lower extremity during the latter stage of the recovery phase, and to predict how performance can be improved. Nine male sprinters were filmed and an inverse dynamic approach was used to determine the lower extremity kinematics and kinetics. Data were averaged to obtain an acceleration-time record of the hip which was representative of all participants. The lower extremity was modeled as two rigid bodies which were connected by a frictionless knee joint and controlled by a resultant muscle torque at the hip and knee. A number of objective functions were selected as criteria to be optimized: (1) minimize the sum of the absolute joint torques, (2) minimize the sum of the absolute muscle powers generated or absorbed at the joints, and (3) repeat the above with an efficiency weighting of 3:1 for negative:positive work. Each optimization criterion was compared to the measured kinematic data obtained from the film analysis to evaluate the accuracy of the simulation. An optimization run was considered to be complete when two conditions were met: the final segment angles were within 0.05 rad and the segmental angular velocities were within 0.5 rad/s of their target values, and a zero gradient had been repeatedly encountered. The results showed the usefulness of applying optimal control theory to sprint running, and indicated that the minimization of the average muscle power generated or absorbed at the hip and knee joints was a possible performance criterion. To minimize hip and knee power required increased knee flexion and higher knee lift than was seen in the experimental results, and suggested that an efficient use of eccentric muscle activity was required. The authors also found that to reducing the leg recovery time was accomplished by reducing knee lift, however, a greater eccentric knee extensor torque was also required in the period just prior to foot strike. Although this research made a significant contribution to the application of optimal control theory to sprinting, they limited their investigation to the portion of the recovery phase from right foot touchdown to left foot touchdown and omitted the initial part of recovery. The portion of the sprint stride immediately after toe-off must be included in the analysis as this position may dictate subsequent leg actions.

There has been a considerable amount of work done on the modeling and simulation of human locomotion, with a few researchers using these methods to investigate sprinting. There are limitations, however, to these studies. The work completed by Philips et al. (1983) and Wood et al. (1987) focused on the recovery phase of sprinting, but neither examined the recovery phase in full from toe-off to ground contact. The initial portion of the recovery phase is important as it may dictate the subsequent kinematics and kinetics of the recovery leg. In addition, the models used in these studies did not include a foot segment; the mass was either being neglected or added to the shank. Philips et al. (1983) felt that since the movement of the foot was minimal during the swing phase, combining the mass of the foot and shank into one segment would not be a limitation on the model. Not including the foot segment in the model would alter the inertial characteristics of the lower leg and could potentially result in inaccurate kinematic estimates from the model.

2.5 Development of Research Questions

The recovery phase of sprinting has been shown on one subject to be the rate limiting factor (Chapman & Caldwell, 1983), therefore good recovery mechanics should be the goal of all sprinters, particularly those at the elite level. This includes a lack of full knee extension at takeoff (Mann, 1985), good knee flexion and a high knee lift (Sinning & Forsyth, 1970), a rapid foot descent with the foot moving backwards relative to the body as it strikes the ground (Mann, 1985), and contact made at the point almost beneath the body's centre of gravity (Kunz & Kaufman, 1981; Mann, 1985). Results from studies involving elite sprinters (Mann & Herman, 1985; Kersting, 1999; Kivi, 1999), however, have shown that sprinting speed does not appear to be dependent on a particular recovery phase movement pattern, nor does there appear to be one universally accepted means of recovering the leg forward.

Mann and Herman (1985) compared kinematics of the first, second, and eighth place finishers in the men's 200 metres at the 1984 Olympics. The purpose of their paper was to compare the kinematics and attempt to identify why the gold medallist outperformed the others. This paper, however, also showed that there are differences in the sprint kinematics among the top 8 sprinters in the world. Tables 2-1 and 2-2 show some of the positional differences found at the hip and knee. In Table 2-1 hip angles were measured between the thigh and trunk with values less than 180° indicating hip extension and greater than 180° hip flexion. The angles seen at the hip were similar for all three subjects at takeoff and full flexion, but the eighth place finisher was much more extended at the hip at full extension than the other two.

Table 2-1. Hip angles reported by Mann and Herman (1985).

Sprinter	Takeoff (deg)	Full Extension (deg)	Full Flexion (deg)
First	167	165	237
Second	170	168	235
Eighth	167	158	239

Knee angle was measured between the thigh and shank with smaller angles indicating more knee flexion (Table 2-2). The recovery leg for the first place finisher showed a full knee flexion position, and maintained this angle through to ankle cross when the ankle of the swing leg passed by the support leg. The second place finisher did not flex the knee to the same degree and was also more extended at ankle cross. The eighth place finisher had the smallest maximum angle of knee flexion but was also more extended at ankle cross.

Table 2-2. Knee angles reported by Mann and Herman (1985).

Sprinter	Takeoff (deg)	Full Flexion (deg)	Ankle Cross (deg)
First	157	38	44
Second	156	43	54
Eighth	158	37	50

The eighth place finisher was more extended at the hip at full extension, and then fully flexed the knee in order to reduce the moment of inertia of the leg to allow it to swing forwards properly. The second place finisher was less extended at the hip at takeoff and full extension, and therefore did not flex the knee to the same extent.

Kersting (1999) reported sprint kinematics for six semi-finalists from the men's 100 metres at the 1997 IAAF World Championships in Athletics, measured at the 70 metre mark of the race. The results are seen in Figure 2-14. Through the recovery phase, differences were seen at the knee angle at touchdown, hip extension velocity at touchdown, horizontal velocity of the foot prior to touchdown, trunk angle at touchdown (measured as the angle between the trunk and the horizontal, and maximum hip flexion angle (measured as the angle between the thigh and the horizontal). It is difficult to draw any conclusions when comparing each individual's kinematics to the final results of the race, as each sprinter showed a unique movement pattern. But considering these athletes were among the top 16 sprinters in the world at the time, there is considerable disparity among them.

Parameter	Ezinwa 10.15	Boldon 10.00	Greene 9.90	Fredericks 9.93	Bailey 9.91	Montgomery 10.08	Mean 9.995	SD 0.010
γ_TD [°]	155.5	155.3	153.0	140.9	155.2	145.5	150.89	6.18
$\Delta\gamma$_TD [°]	20.6	22.3	11.4	5.3	14.4	9.8	13.98	6.52
$d\gamma/dt_{mean}$ [°/s]	415.0	371.3	324.0	235.3	189.3	351.5	314.40	85.69
$d\gamma/dt_{max}$ [°/s]	597.0	542.7	439.4	346.8	326.6	211.5	410.70	144.06
$d\beta/dt$_TD [°/s]	491.3	362.0	170.9	519.5	558.0	695.0	465.59	180.56
v_x Tip [m/s]	2.58	2.70	2.43	1.48	2.18	1.40	2.13	0.56
τ_TD [°]	78.8	87.1	77.0	84.1	85.6	80.2	82.16	4.04
δ_{min} [°]	19.9	13.7	17.1	9.3	20.5	17.4	16.25	4.14

γ_TD [°]	knee angle at touch-down (TD)
$\Delta\gamma$_TD [°]	knee flexion during stance
$d\gamma/dt_{mean}$ [°/s]	mean velocity of knee flexion
$d\gamma/dt_{max}$ [°/s]	maximum instantaneous knee flexion velocity
$d\beta/dt$_TD [°/s]	hip extension velocity at TD
v_x Tip [m/s]	horizontal velocity of the foot tip prior to TD
τ_TD [°]	trunk angle at TD
δ_{min} [°]	maximum knee lift during swing

Figure 2-14. Kinematic data for six semi-finalists from the men's 100 metres at the 1997 IAAF World Championships in Athletics (Kersting, 1999).

A third study which shows variation among elite sprinters is a kinematic analysis of Donovan Bailey (Kivi, 1999). At the time of the research project, Bailey was the world record holder in the 100 metres at 9.84 seconds. Video was collected at the 70 metre mark of the men's 100 metre final at the 1998 Canadian National Track and Field Championships. The results showed that there were differences between Bailey's sprint kinematics and those reported in the literature on other elite sprinters, with the most dramatic differences seen at the knee. Bailey flexed the knee less during the recovery phase to an angle of

47°. Sprinters should flex the knee fully to decrease the moment of inertia of the recovery leg; however, for Bailey this was not the case. As he recovered his leg forward, he reached a maximum knee extension angle of 159° before extending at the hip and moving his foot to the ground for support. In comparison, a knee extension angle of 150° has been reported by Mann (1985) for elite sprinters.

From these studies, it is apparent that sprinting speed is not dependent on one particular recovery phase movement pattern. These studies all involved elite athletes, yet differences were seen in a number of variables throughout the recovery phase, from takeoff to touchdown. Despite these differences, the movement of the recovery leg is governed by the interaction of these variables (Mena, Mansour, & Simon, 1981). This suggests that even though there may be variations in the sprint mechanics among elite athletes, relationships exist among the variables which define and control the movement.

In reviewing previously published literature on the biomechanics of elite sprinting, there are two important factors regarding the recovery phase that have not been considered or examined previously. First, despite the fact that there is biomechanical data showing that sprinting speed is not dependent on a particular recovery phase movement pattern, there have been no studies which have examined and quantified the relationships among the variables which govern the motion of the recovery leg among elite sprinters. The three studies discussed above, for example, were descriptive in nature and based any conclusions on the observed similarities or differences among the participants. Second, there has been no research which has considered the kinetic differences among the sprinters and their relationships with the kinematics of motion. Kinematic analysis simply describes the motion without providing insight into what causes a body to move how it does; kinetic studies are those which deal with the forces producing movement (Hay, 1993). Previous studies which have considered the relationships among the kinematics and kinetics of sprinting have only examined the correlations between sprinting speed and muscle strength (Alexander, 1989) or ground reaction force production (Mero & Komi, 1986). In order to understand the differences among sprinters, it is necessary to consider both the kinetic and kinematic variables that interact to produce the resulting recovery leg movements. These include the position of the leg at takeoff, the angular displacements and angular velocities of the limb segments, as well as the muscle moments controlling the motion at each joint. The way in which these variables interact has not been investigated. If sprinters do not all recover the leg in the same manner, there must be some systematic way that these kinematic and kinetic factors work together to produce the resulting movement pattern. Therefore, one of the purposes of this study is to determine what are the relationships among the kinetic and kinematic variables of the recovery phase of elite sprinters.

It is important to understand these kinematic and kinetic relationships in order to explain why there are differences seen among individuals in the manner with which they recover their leg forwards. This interaction would play a vital role in identifying the strengths and weaknesses of each individual athlete, which could influence the training methods used by each to improve performance.

Based on the results of previous sprint literature, it is hypothesized that there are relationships among these variables which will determine the sequence of movements from takeoff to touchdown. Sprinters with greater extension at the hip and knee at takeoff will also show greater knee flexion during recovery and faster hip flexion angular velocity with an increase in hip flexion torque. This in turn will result in greater peak hip flexion angles and faster peak hip extension angular velocities. Sprinters who limit their hip and knee extension will not flex the knee to the same extent, will have slower hip flexion angular velocities and will have greater hip flexion torque. They will also show smaller peak hip flexion angles and slower hip extension angular velocities.

If a number of variables interact to produce the motion of the recovery leg, it is reasonable that the kinematics of the leg at any point is dictated by preceding positions and movements. For a given action or skill, however, the motion can vary greatly depending on the initial position of the body (Chou, Song, & Draganich, 1995). This means that the position of the leg at takeoff during sprinting is important in determining the overall movement pattern seen during recovery.

Previous research has shown that sprinters who are more extended at both the hip and knee at takeoff show greater knee flexion through recovery in order to reduce the moment of inertia and allow the leg to swing through adequately. Conversely, sprinters who are less extended at the hip and knee do not require as much knee flexion during recovery (Kivi, Maraj, & Gervais, 2002b). These results were based on comparing two groups of athletes of different sprinting abilities, but they suggest that takeoff position plays a role in determining various other factors such as the degree of knee flexion or the angular velocities of the thigh or shank.

Better sprinters tend to minimize the amount of extension at both the hip and knee at the point of takeoff in order to minimize ground contact time (Mann, 1985), therefore it may be advantageous for an athlete to make modifications in technique in order to improve performance. There has been no research, however, that has examined the influence of changing these takeoff positions on the subsequent movements of the recovery leg. In order to perform this type of analysis it would be necessary to have a subject modify his/her sprinting mechanics and then quantify the resulting changes in the recovery leg. Performing a study of this nature would likely be very harmful to the individual as there is an increased risk of injury (Vaughan, 1984), and therefore must be completed using computer simulation methods.

Computer simulation of human gait showed no variation from the normal motion when modifications of less than 20° were made to the initial angular conditions at the hip, knee, and ankle (Mena, Mansour, & Simon, 1981). These results are not unexpected, as the joint moments and ranges of motion during gait are considerably smaller than those during sprinting (Novacheck, 1996). In a more dynamic movement such as sprinting, it is likely that the hip and knee angles at takeoff will be more influential on the subsequent recovery leg movements, but these relationships are currently not known. Previous simulation studies investigating the movements of the leg during the recovery

(Wood et al., 1987; Philips et al., 1983) have not recognized the potential importance of this aspect of the recovery phase and omitted it in their analyses. Therefore, the second objective of this study is to use computer simulation methods to determine what effect changing hip and knee angles at takeoff would have on the subsequent kinematics of the recovery leg.

This type of analysis would further explain the relationship among the kinematic and kinetic variables associated with the recovery phase of sprinting, as changing one aspect of the movement and quantifying the resulting motion would reveal how the variables interact. Based on the findings of Kivi et al. (2002b) and the relevant sprint literature, it is hypothesized that more hip and knee extension at takeoff will lead to small knee flexion angles because the rotational inertia of the leg must be reduced to allow the leg to recover adequately. This reduced moment of inertia will result in faster hip flexion angular velocities and larger angles of hip flexion. Conversely, less extension at the hip and knee at takeoff will result in less knee flexion. This will increase the rotational inertia and cause slower hip flexion angular velocities and smaller angles of hip flexion.

CHAPTER 3

Project 1 – Analysis of Recovery Leg of Elite Sprinters

3.1 Purpose

The purpose of this study was to analyze the kinematics and kinetics of the recovery leg in order to understand the relationships among the variables producing the differences seen in the movement patterns during the recovery phase of elite sprinters. It was hypothesized that there would be statistically significant correlations among the variables which determine the sequence of movements of the lower extremity from takeoff to touchdown. Sprinters with greater extension at the hip and knee at takeoff would also show increased knee flexion during recovery and faster hip flexion angular velocities. This in turn would result in increased hip flexion angles and enable the sprinters to produce faster hip extension angular velocities. Sprinters who limit hip and knee extension at takeoff would not flex the knee to the same extent, resulting in slower hip flexion angular velocities. This would reduce the amount of hip flexion and result in slower hip extension angular velocities.

3.2 Methodology

3.2.1 Subjects and Testing Protocol

Participants for this study were the semi-finalists and finalists from the 2001 Canadian Senior Track and Field Championships and the 8th IAAF World Championships in Athletics, which were held in Edmonton, Alberta, Canada. Subject characteristics are seen in Table 3-1, with data from 14 sprinters being used in the analysis. More subjects would have been preferable; however due to overlap from lane 1 to 8 in each race it was not possible to obtain one full stride from any other individuals. Subjects 1 to 9 were from the semi-finals and finals of the 2001 Canadian Senior Track and Field Championships, whereas subjects 10 to 14 were from the semi-finals and finals of the 2001 IAAF World Championships in Athletics. The mean age ± standard deviation for all subjects was 26.1 ± 4.2 years, height was 1.78 ± 0.03 metres, mass was 76.6 ± 4.6 kilograms, and time in the 100 metre race analyzed was 10.32 ± 0.32 seconds. The Faculty Research Ethics Board approved procedures for the experiment and participant involvement.

Table 3-1. Subject characteristics.

Subject	Age (yrs)	Height (m)	Mass (kg)	100m time (s)
1	21	1.81	73	10.29
2	21	1.80	80	10.29
3	21	1.75	75	10.40
4	26	1.70	75	10.73
5	25	1.80	77	10.83
6	24	1.78	75	10.64
7	34	1.82	83	10.24
8	33	1.80	83	10.48
9	31	1.78	70	10.58
10	23	1.78	83	9.94
11	26	1.80	80	10.31
12	26	1.79	69	9.85
13	28	1.75	75	10.12
14	27	1.76	75	9.82
MEAN	26.1	1.78	76.6	10.32
S.D.	4.2	0.03	4.6	0.32

Two JVC GVL-9800 digital video cameras, each recording at 120 Hz with a shutter speed of 1/250th, were placed at approximately 55 and 60 metres from the start line such that the optical axis of each camera was perpendicular to the plane of motion. The field of view for each camera was 7 metres with an overlap of 2 metres which allowed the researcher to combine the views into one continuous sequence. The cameras were located on the inside of the track, 8 metres from the edge of lane 1, and were mounted on a tripod at a height of approximately 1.0 metre from the ground.

Early in the morning prior to the day's competition, the cameras were set up and the field of view was calibrated. Two vertical rods with points of known distances were placed in the field of view for each camera and recorded. The rods were a standard distance apart, and were positioned in each lane to allow for analysis of data in all eight lanes.

The JVC GVL-9800 digital video cameras are capable of capturing at 120 Hz; however, there is a tradeoff between sampling frequency and image resolution. When the “high speed” mode is selected during recording, the full screen (720 x 480) at 60 Hz is split horizontally into two images, resulting in a lower resolution of 720 x 240 at 120 Hz (Ariel Dynamics, 2006). Sacrificing image resolution for higher sampling rates was necessary for this investigation as the higher sampling rate helped with the identification of specific events in the sprint stride such as take-off and touchdown, to prevent aliasing errors, and to reduce the amplification of the noise in the signal when the positional data was differentiated twice to calculate acceleration (Lanshammar, 1982).

3.2.2 Data Reduction and Analysis

Data processing was completed using the Ariel Performance Analysis System (APAS). 2D reconstruction of the raw positional data was accomplished using a modified Direct Linear Transformation technique. The time-dependent coordinates of each landmark were smoothed using a low-pass digital filter with an appropriate cutoff frequency to reduce small random errors that may have occurred during digitizing. The cutoff frequency was determined by inspection of the raw and filtered data and comparison between the respective power spectra and ranged between 5 and 8 Hz. A screen capture of the APAS data filtering module is seen in Figure 3-1.

A four linked rigid body segment model requiring six segmental endpoints was devised to allow for analysis of the leg throughout the entire sprint cycle. The anatomical landmarks were the shoulder (tip of acromion), hip (greater trochanter), knee (lateral epicondyle), ankle (lateral malleolus), heel (middle of calcaneus), and toe (base of fifth metatarsal) on the left side of the body. For each subject one complete stride was analyzed, with a stride defined as ground contact of one foot to the subsequent ground contact of the same foot. Central differences were used to calculate velocity and acceleration data for each joint. In addition, angular position, velocity, and acceleration data were calculated for each segment. Stance and flight times were measured in video frames and converted to time, with stride frequency calculated as the reciprocal of the sum of the stance and flight times.

In order to determine the net joint forces and moments acting at the hip, knee, and ankle, an inverse dynamics approach was used. A planar model was used to simulate the recovery phase of sprinting. The model was similar to that

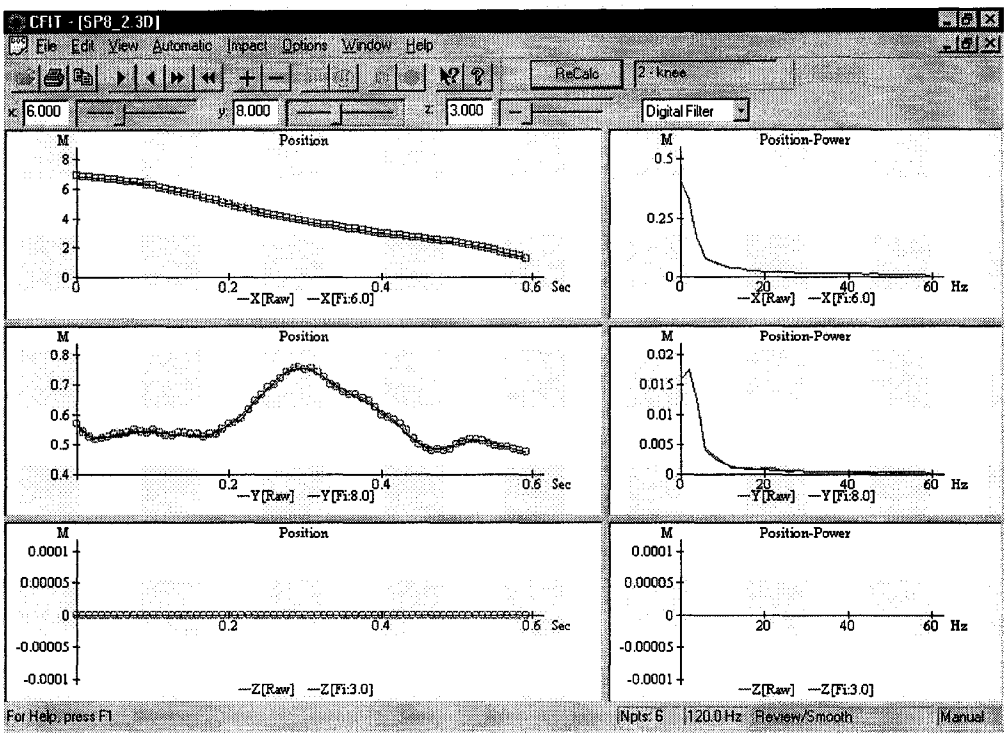


Figure 3-1. Screen capture of the APAS data filtering module, showing the x and y positional data (left) and the respective power spectrum (right).

of Wood et al. (1987) with three segments: the foot (segment 1), shank (segment 2), and thigh (segment 3) (Figure 3-2). Wood et al.'s model consisted of only two segments, as the foot was omitted. To describe the orientation and configuration of the model, the thigh angle λ was defined as the angle between the thigh segment and the right horizontal. The shank angle θ was defined as the angle between the shank and the right horizontal, and the foot angle ϕ was defined as the angle between the top of the foot segment and the left horizontal (Figure 3-2). The inertial reference frame R:OXYZ was oriented such that the Y axis is directed vertically upwards. This analysis was restricted to two-dimensional motion in the vertical XY plane of the inertial reference frame R:OXYZ. This assumption reduced the mathematical complexity of the problem without affecting the generality of the solution procedure, as a sagittal plane analysis can identify the majority of the important details about running (Robertson & Caldwell, 2004). The origin was set as an unobstructed fixed point within the field of view.

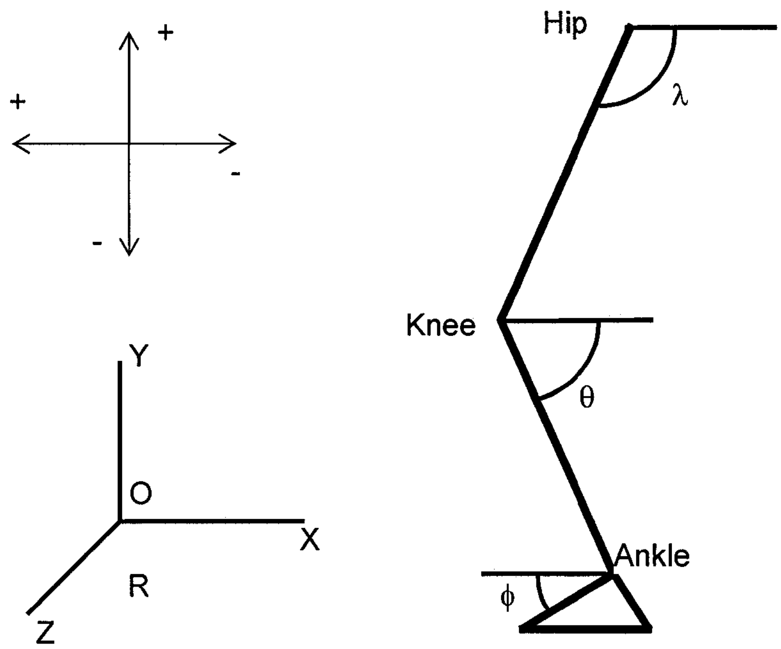


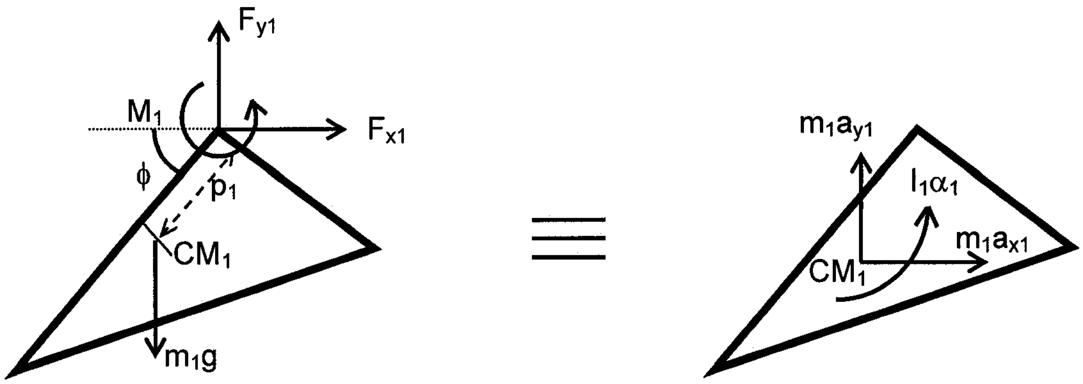
Figure 3-2. Orientation of planar model used to simulate the recovery phase of sprinting.

3.2.3 Equations of Motion

A Newtonian approach was used to derive the equations of motion, which states the sum of the forces acting on a segment are equal to the mass times the acceleration, and the sum of the moments acting on a segment are equal to the moment of inertia times the angular acceleration for linear and rotational movement, respectively. These equations were adapted from Dillman (1971) in his kinetic analysis of the recovery leg during sprinting.

$$\Sigma F = ma \quad \text{and} \quad \Sigma M = I\alpha$$

Segment 1 - Foot

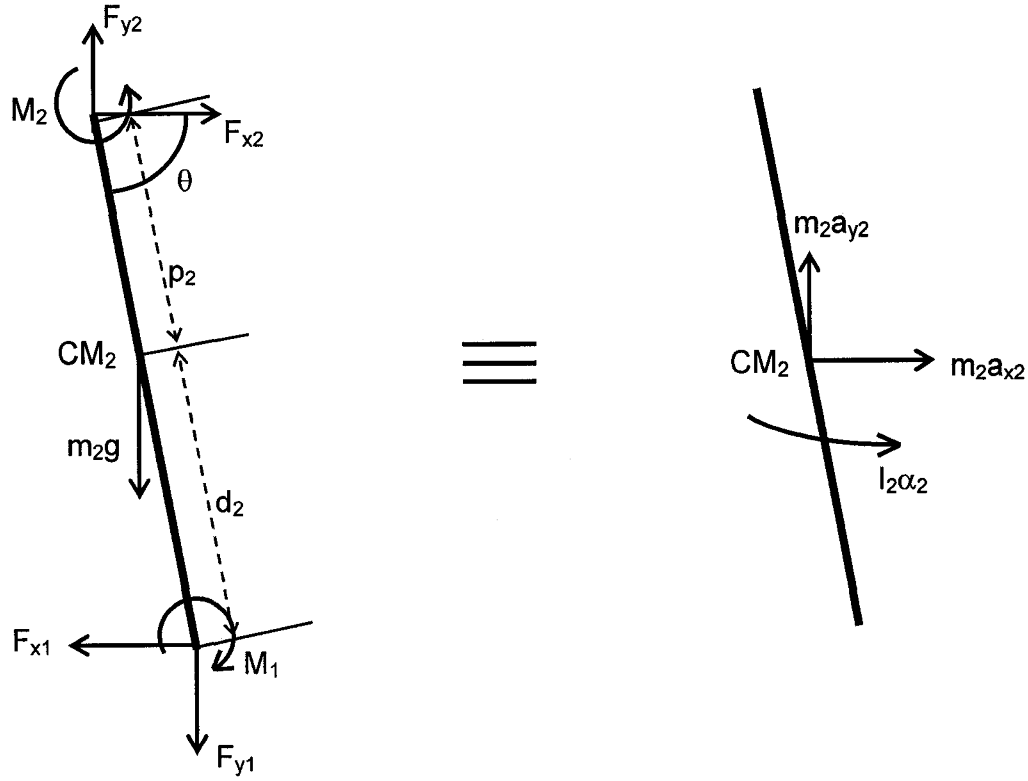


$\Sigma F_x = ma_x$
$F_{x1} = m_1 a_{x1}$ (1)

$\Sigma F_y = ma_y$
$F_{y1} - m_1 g = m_1 a_{y1}$
$F_{y1} = m_1 a_{y1} + m_1 g$ (2)

$\Sigma M_1 = I_1 \alpha_1$
$M_1 - F_{x1} * FA_{x1} + F_{y1} * FA_{y1} = I_1 \alpha_1$
$M_1 - (F_{x1} * (p_1 * \sin \phi)) + (F_{y1} * (p_1 * \cos \phi)) = I_1 \alpha_1$ (3)

Segment 2 - Shank

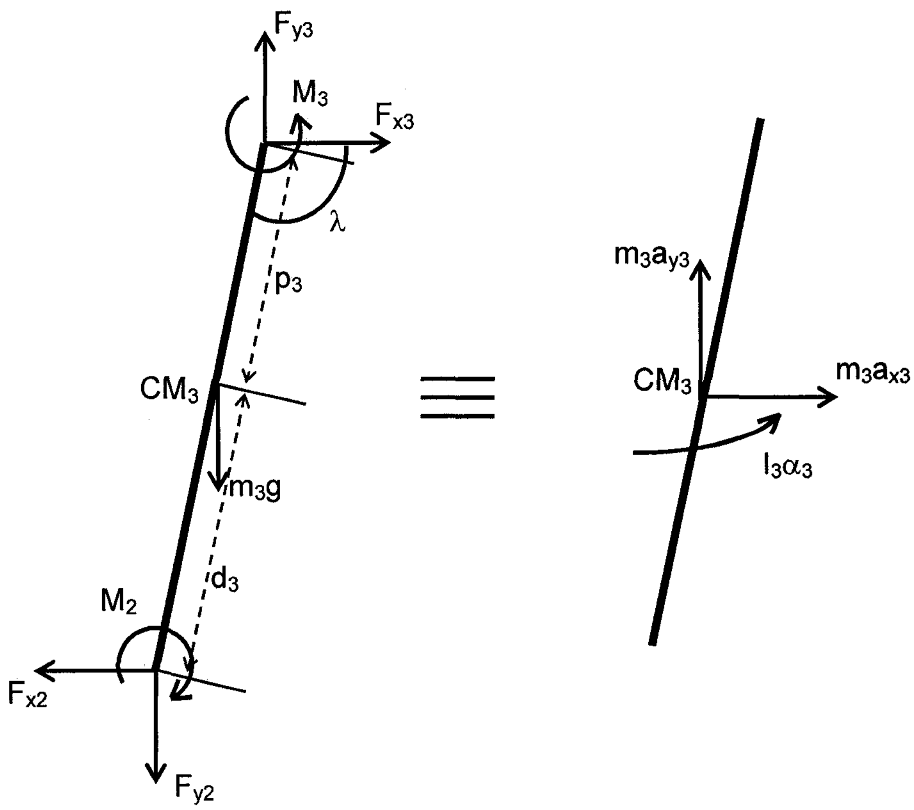


$\Sigma F_x = ma_x$
$Fx_2 - F_{x1} = m_2a_{x2}$
$Fx_2 = m_2a_{x2} + F_{x1}$ (4)

$\Sigma F_y = ma_y$
$F_{y2} - F_{y1} - m_2g = m_2a_{y2}$
$F_{y2} = m_2a_{y2} + F_{y1} + m_2g$ (5)

$\Sigma M_2 = I_2\alpha_2$
$M_2 - M_1 - F_{x2}{*}FA_{x2} - F_{y2}{*}FA_{y2} - F_{x1}{*}FA_{x1} - F_{y1}{*}FA_{y1} = I_2\alpha_2$
$M_2 - M_1 - (Fx_2{*}(p_2{*}\sin\theta)) - (Fy_2{*}(p_2{*}\cos\theta)) - (F_{x1}{*}(d_2{*}\sin\theta))$
$- (F_{y1}{*}(d_2{*}\cos\theta)) = I_2\alpha_2$ (6)

Segment 3 - Thigh



$\Sigma F_x = ma_x$
$F_{x3} - Fx_2 = m_3a_x$
$F_{x3} = m_3a_x + F_{x2}$ (7)

$\Sigma F_y = ma_y$
$F_{y3} - F_{y2} - m_3g = m_3a_y$
$F_{y3} = m_3a_y + F_{y2} + m_3g$ (8)

$\Sigma M_3 = I_3\alpha_3$
$M_3 - M_2 - Fx_3{*}FA_{x3} + F_{y3}{*}FA_{y3} - F_{x2}{*}FA_{x2} + F_{y2}{*}FA_{y2} = I_3\alpha_3$
$M_3 - M_2 - (Fx_3{*}(p_3{*}\sin\lambda)) - (Fy_3{*}(p_3{*}\cos\lambda)) - (F_{x2}{*}(d_3{*}\sin\lambda))$
$- (F_{y2}{*}(d_3{*}\cos\lambda)) = I_3\alpha_3$ (9)

Definition of Variables:

F_{x1}, F_{x2}, F_{x3} = x-component of the joint reaction force at joints 1, 2, and 3

F_{y1}, F_{y2}, F_{y3} = y-component of the joint reaction force at joints 1, 2, and 3

M_1, M_2, M_3 = resultant joint moment at joints 1, 2, and 3

CM_1, CM_2, CM_3 = centre of mass of segments 1, 2, and 3

I_1, I_2, I_3 = moment of inertia of segments 1, 2, and 3

α_1, α_2, α_3 = angular acceleration of segments 1, 2, and 3

m_1, m_2, m_3 = mass of segments 1, 2, and 3

g = acceleration due to gravity

p_1, p_2, p_3 = proximal distance of the CM for segments 1, 2, and 3

d_2, d_3 = distal distance of the CM for segment 1, 2, and 3

a_{x1}, a_{x2}, a_{x3} = linear acceleration in the x-direction of the centre of mass of segment 1, 2, and 3

a_{y1}, a_{y2}, a_{y3} = linear acceleration in the y-direction of the centre of mass of segment 1, 2, and 3

The location of the mass centre of each segment was relative to the origin of O of R:OXYZ was specified by (x_s, y_s) and the proximal-distal orientation of the long axis of each segment *i* relative to the X axis of R was specified by angle λ (thigh), θ (shank), and ϕ (foot). Furthermore, the moment of inertia of segment *s* about a transverse Z axis passing through its centre of mass was denoted by I_s.

Hip, knee and ankle resultant joint moments were calculated with a custom program written in Matlab (Natick, MA: The Math Works, Inc.). In order to verify the accuracy of the calculations, five frames were randomly selected among the 14 subjects and at different points in time through the recovery phase and the joint moments acting at the ankle, hip, and knee were calculated manually using equations (3), (6), and (9). The joint moments calculated manually were the same as those output from the Matlab program.

3.2.4 Body Segment Parameter Data

To describe and quantify the inertial properties for the sprinters in the mathematical model, the body segment parameter data from Zatsiorski and Seluyanov (1983) as modified by DeLeva (1996) was used. These data consist of the percentage mass of the three segments, the segments' centre of mass locations expressed as a percent of the total length of the segment, and the radius of gyration expressed as a percent of the total length of the segment. The height and weight for the subjects were used to determine the inertial parameters for each individual and were obtained from athlete biographies on the IAAF website (IAAF, 2001) and from the Athletics Canada website (Athletics Canada, 2001).

3.2.5 Statistical Analysis

Descriptive statistics (means (M) and standard deviations (SD)) were calculated for variables selected for analysis based on previous sprint studies (Kivi et al., 2002; Hoskisson & Korchemny, 1991; Mann, 1985; Mann & Herman; 1985; Mann & Sprague, 1983; Mann, 1981). Pearson product-moment correlation coefficients were calculated to determine where relationships existed among the kinematic and kinetic variables which interact to govern the movement of the recovery leg during sprinting (Mena, Mansour, & Simon, 1981), and to establish the strength of these relationships. The kinematic and kinetic variables included in the analysis were defined as follows:

Stride Length – the horizontal distance between the termination of ground contact of one foot to the subsequent ground contact of the same foot

Stride Rate – the number of strides completed per second

Hip angle at takeoff – the absolute angle of the hip, as measured between the right horizontal and the thigh segment, at the moment of takeoff

Maximum angle of hip extension – the smallest numerical value, which represents the largest angle of hip extension, measured as an absolute angle between the right horizontal and the thigh segment after takeoff

Knee angle at takeoff – the relative angle of the knee, as measured between the thigh and shank segments, at the moment of takeoff. Full knee extension was 180 degrees, with smaller angles representing increased knee flexion

Minimum knee angle – the smallest relative angle measurement at the knee between the thigh and shank segments during leg recovery

Minimum moment of inertia of the recovery leg – is the smallest value calculated for the moment of inertia of the recovery leg during the recovery phase

Maximum hip flexion angle – the largest numerical value representing the largest angle of hip flexion during leg recovery, measured as an absolute angle between the right horizontal and the thigh segment,

Horizontal foot velocity prior to touchdown – the horizontal velocity of the foot immediately prior to contact with the ground

Peak hip flexion angular velocity – the fastest flexion angular velocity measured at the hip during leg recovery

Peak hip extension angular velocity – the fastest extension angular velocity measured at the hip during leg recovery

Peak knee flexion angular velocity – the fastest flexion angular velocity measured at the knee during leg recovery

Peak knee extension angular velocity – the fastest extension angular velocity measured at the knee during leg recovery

Peak hip flexor moment – the largest flexor moment measured at the hip during leg recovery

Peak hip extensor moment – the largest extensor moment measured at the hip during leg recovery

Peak knee flexor moment – the largest flexor moment measured at the knee during leg recovery

Peak knee extensor moment – the largest extensor moment measured at the knee during leg recovery

3.2.6 Reconstruction Accuracy and Error Analysis

To estimate the reconstruction accuracy and the potential systematic bias introduced to the data, the locations of the digitized calibration points were reconstructed. Average estimates of the coordinate errors of the reconstructed locations of the calibration markers were found to be 0.012 metres and 0.013 metres in the x and y directions, respectively.

Because the video collected for analysis was of elite athletes during competition, there was no opportunity to measure segment lengths directly from the participants or to attach markers to define segmental endpoints, which would result in some digitizing error introduced by the manual digitization process. In order to estimate the amount of random error introduced and to determine the reliability of the data, a digitization test-retest analysis was completed. One trial was randomly selected and digitized a second time, and segmental x and y coordinates compared between trials. The criterion used to estimate the error between trials was the root mean square (RMS) error. The results of the test-retest analysis are seen in Table 3-2. The poorest precision estimates were found at the hip, with better results seen at the knee and ankle. This is likely because the smaller size of the knee and ankle joint areas provide for more precise estimation of the joint centre.

Table 3-2. Average digitizing precision for lower extremity landmarks.

	Hip		Knee		Ankle	
	X	Y	X	Y	X	Y
RMS Error	0.013 m	0.014 m	0.009 m	0.010 m	0.008 m	0.010 m

A second test was completed to estimate the precision of angle calculations. A different trial was randomly selected by the investigator and digitized a second time, with hip (λ) and knee (θ) angles compared between trials. The RMS precision estimates were 1.31° for the hip and 0.58° for the knee. Table 3-3 presents the average RMS error estimates between the raw and filtered data for the x and y coordinates for the digitized body landmarks for all participants. The larger RMS error estimates for the hip were a result of the reduced precision in estimating the joint centre.

Table 3-3. RMS error estimates of the coordinate data.

	X Error (m)	Y Error (m)
Shoulder	0.011	0.012
Hip	0.013	0.014
Knee	0.008	0.010
Ankle	0.010	0.009
Heel	0.011	0.010
Toe	0.009	0.010

3.3 Results and Discussion

3.3.1 Stride Characteristics

A mean stride length of 4.90 ± 0.18 metres was found for the subjects (Table 3-4), with a mean stride rate of 2.38 ± 0.07 strides/s. Mean contact time and flight time were 0.09 ± 0.01 seconds and 0.11 ± 0.01 seconds, respectively.

Table 3-4. Stride characteristics.

	Stride Length (m)	Stride Rate (strides/s)	Contact Time (s)	Flight Time (s)
Mean	4.90	2.38	0.09	0.11
S.D.	0.18	0.07	0.01	0.01

Stride characteristics were similar to those of previous sprint biomechanics literature involving national and international level sprinters. Nummela, Vuorimaa, and Rusko (1992) reported stride lengths of 5.00 m in national level sprinters, and Ae, Ito, and Suzuki (1992) found stride lengths ranging from 4.70 to 5.04 m and stride rates between 2.27 and 2.48 strides/s at the 70 metre mark in the finalists of the men's 100 metre final at the 1991 World Championships in Athletics. Contact and flight times were comparable to those found in other studies involving elite sprinters. Contact times of 0.09 seconds have been reported by Mero, Luhtanen and Komi (1986), and 0.10 seconds were found by Burt (1994) in studies of elite sprinters. Mann (1985) reported "good" elite male sprinters achieve a stride rate of 2.4 strides/s, while "average" elite male sprinters achieve 2.25 strides/s and "poor" elite male sprinters achieve 2.1 strides/s.

3.3.2 Kinematics and Kinetics of the Hip

Sprint-specific kinematic and kinetic variables at the hip are presented in Table 3-5 for all subjects. Mean hip extension angle at take-off was 73 ± 4°, with a mean maximum angle of hip extension of 66 ± 4°. The mean maximum hip flexion angle seen through recovery was 155 ± 4°. Mean hip flexion and extension angular velocities of –729 ± 92°/s and 477 ± 57°/s, respectively, were found for the subjects. Peak hip extensor moment through recovery was 266 ± 22 Nm, and peak hip flexor moment was –276 ± 23 Nm.

Table 3-5. Kinematics and kinetics of the hip.

	Hip Angle at Takeoff (°)	Maximum Hip Extension (°)	Maximum Hip Flexion (°)	Hip Flexion Angular Velocity (°/s)	Hip Extension Angular Velocity (°/s)	Hip Extensor Moment (Nm)	Hip Flexor Moment (Nm)
Mean	73	66	155	-729	477	266	-276
S.D.	4	4	4	92	57	22	23

Figures 3-3 shows the mean joint moment, angular acceleration, angular velocity, and angular displacement through the recovery phase for all subjects. At takeoff, there was a dominant flexor muscle moment which acted to slow the backwards rotation of the thigh and initiate the forwards swing of the recovery leg. As the thigh passed the body the net joint moment shifted to that of the

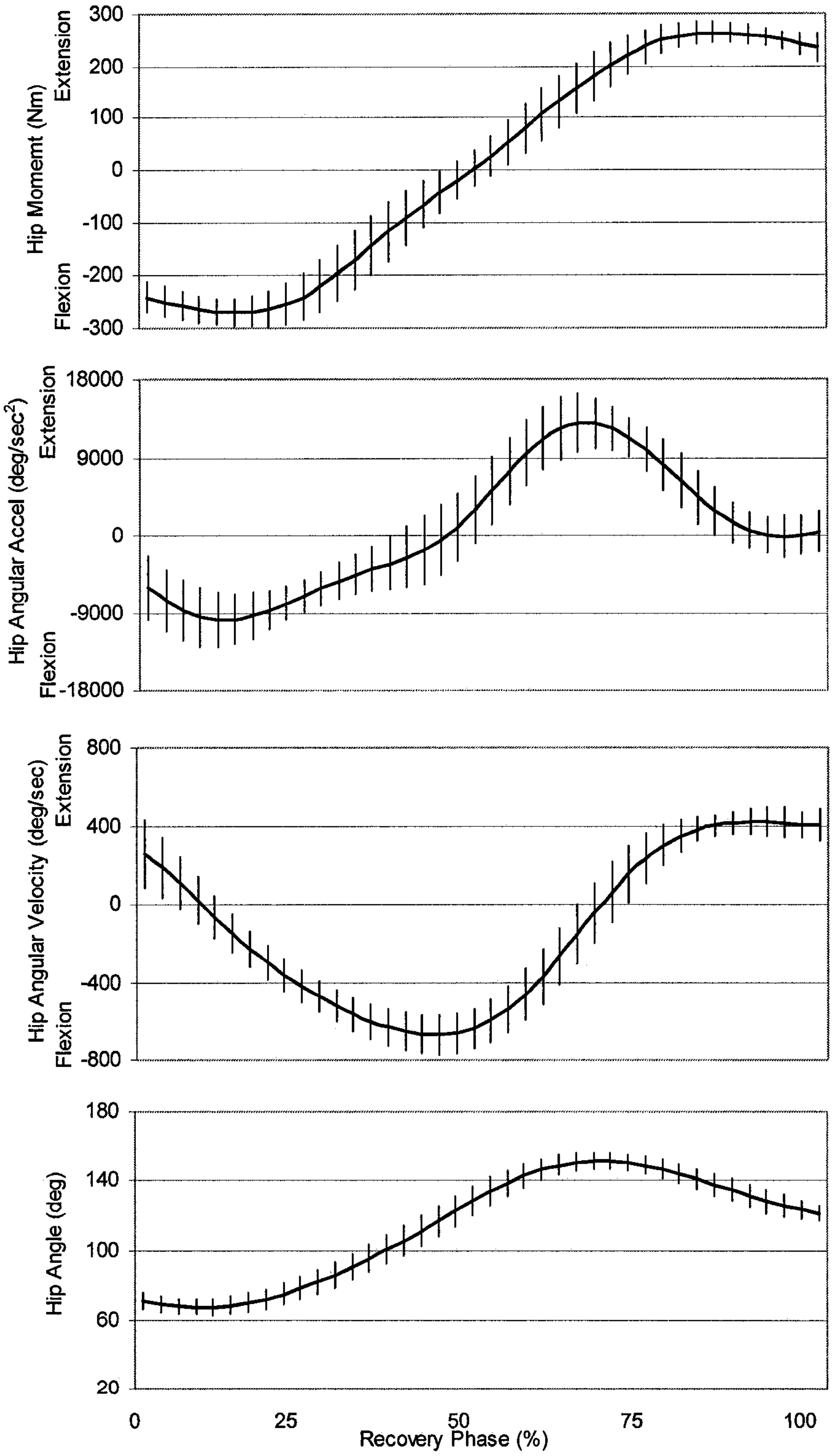


Figure 3-3. Average hip joint moment, angular acceleration, angular velocity, and hip joint angle during the recovery phase of sprinting across all participants. Vertical bars represent standard deviations.

extensors, which slowed the forward rotation of the thigh and initiated the backwards rotation of the leg prior to ground contact.

3.3.3 Kinematics and Kinetics of the Knee

Table 3-6 shows the kinematics and kinetics of the knee for all subjects, including mean and standard deviation values. At take-off, mean knee extension angle was 146 ± 6°, while the minimum knee flexion angle through recovery was 42 ± 7°. Mean knee flexion and extension angular velocities were 1134 ± 162°/s and –1066 ± 141°/s, respectively. At the knee, the peak extensor moment was –116 ± 12 Nm and the peak flexor moment was 142 ± 29 Nm.

Table 3-6. Kinematics and kinetics of the knee.

	Knee Angle at Takeoff (°)	Minimum Knee Angle (°)	Knee Flexion Angular Velocity (°/s)	Knee Extension Angular Velocity (°/s)	Knee Extensor Moment (Nm)	Knee Flexor Moment (Nm)
Mean	146	42	1134	-1066	-116	142
S.D.	6	6	162	141	12	29

The mean knee joint moment, angular acceleration, angular velocity, and angular displacement during the recovery phase for all subjects is seen in Figure 3-4. The net joint moment at the knee at takeoff was dominated by the extensors which functioned to stop the backwards rotation of the shank. Once knee flexion had been halted, the net joint moment became one of extension as the shank was rotated anteriorly. After reaching maximum knee extension angular velocity, the knee flexor moment slowed the forward rotation of the shank. This was followed by a small degree of knee flexion in preparation for ground contact.

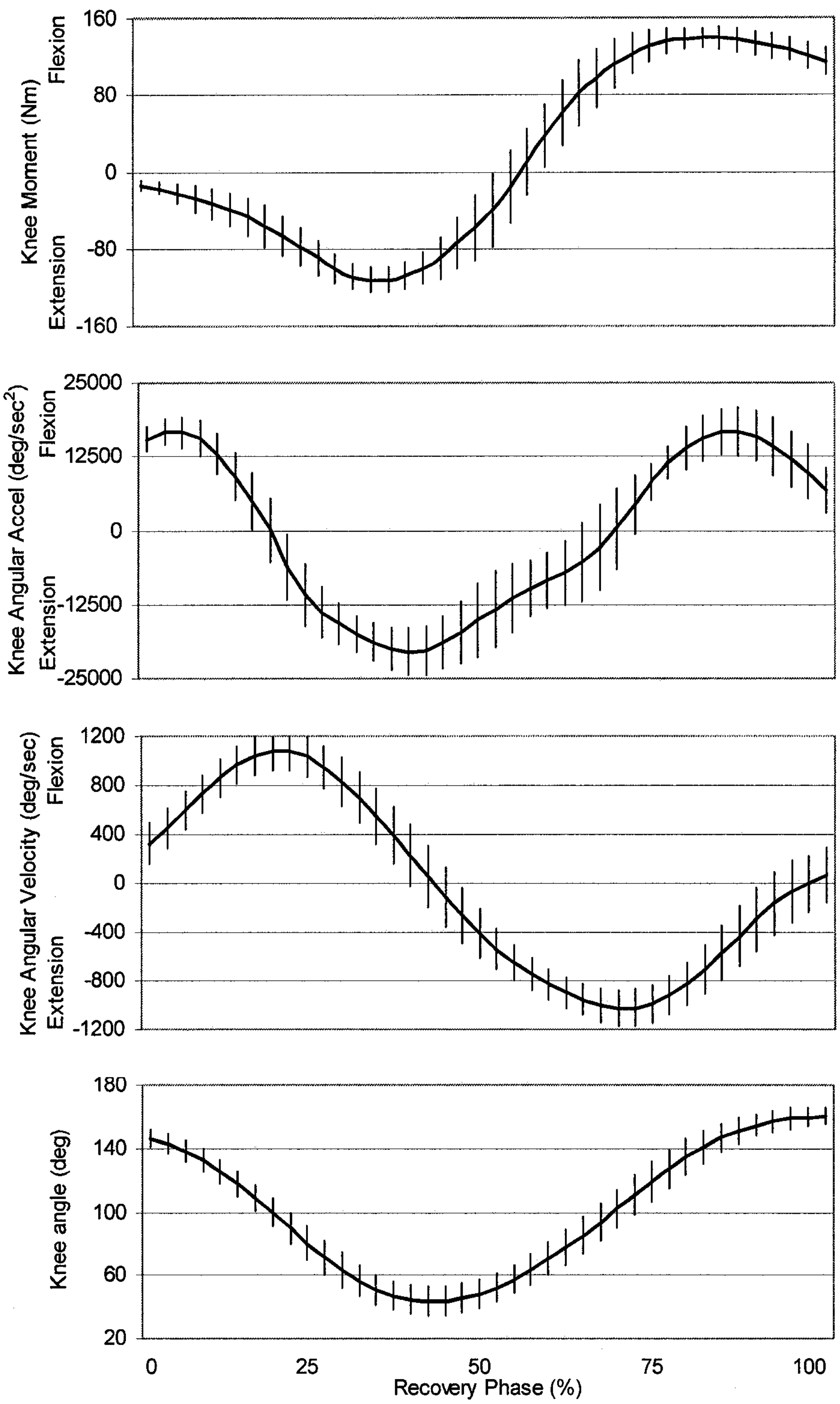


Figure 3-4. Average knee joint moment, angular acceleration, angular velocity, and knee joint angle during the recovery phase of sprinting across all subjects. Vertical bars represent standard deviation values.

An angle-angle diagram is the plot of one joint as a function of another angle at equal intervals in time. That is, one joint angle is used for the abscissa and the other is for the ordinate. For the angle-angle diagram to be meaningful, a functional relationship should exist between the two angles, such as a comparison of hip and knee. According to Grieve (1968), angle-angle diagrams emphasize the relationships between angles more clearly than with separate plots. Milner et al. (1973) felt the angle-angle diagram can be a useful method of presenting data because of the considerable amount of information they convey very simply and in view of the distinct patterns obtained from the subjects tested.

Figure 3-5 shows the hip and knee angle-angle diagrams for all participants. Starting at takeoff (Point 1), there was an initial rapid knee flexion with a small amount of hip flexion. As the knee reached a minimum angle (Point 2) there was a rapid hip flexion. This was followed by an extension of the knee while approaching maximum hip flexion (Point 3) and the foot was positioned for the subsequent ground contact (Point 4).

Further analysis of these graphs reveals additional information regarding the movement of the recovery leg during sprinting. Hip and knee angles at takeoff varied among the athletes, and differences were seen in the minimum knee angle. In addition, there was some variation in the hip angle which corresponded with the minimum knee angle. Knee extension appeared to be more consistent as the graph lines showing less dispersion. Hip and knee angles at touchdown also showed disparity among the athletes.

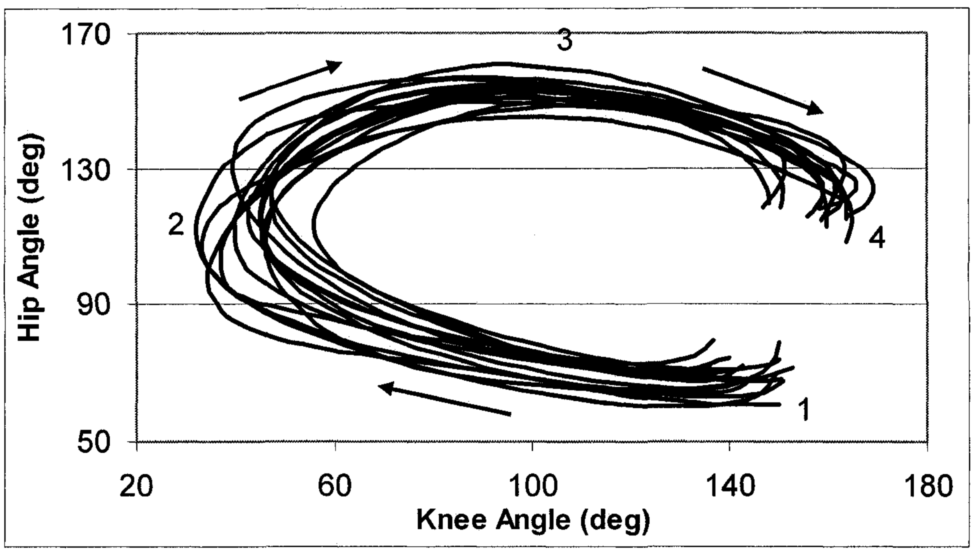


Figure 3-5. Angle-angle diagram of the hip and knee for all participants (n = 14) during the recovery phase. Point 1 = takeoff; Point 2 = minimum knee angle; Point 3 = maximum hip flexion; Point 4 = touchdown.

3.3.4 Correlational Analysis

When the Pearson product-moment correlation coefficients were calculated for the kinematic and kinetic variables of interest, a number of statistically significant relationships were found throughout the recovery phase. Table 3-7 presents the results of the correlational analysis. To facilitate the discussion of the correlational analysis, the recovery phase will be divided into 3 parts: early recovery which is the period immediately after takeoff and includes initial knee flexion; middle recovery when the leg moves from behind the body to in front and is when the minimum knee angle is seen; and late recovery when the forward swing of the leg is stopped and the leg prepares for the subsequent ground contact. In addition, the correlations between both stride rate and stride length and the other sprint variables will be discussed.

Table 3-7. Correlation matrix for the kinematic and kinetic variables related to the recovery phase. Statistically significant correlations are highlighted ($p < 0.05$).

	Hip Angle at TO	Knee Angle at TO	Max Angle Hip Extension	Minimum Knee Angle	Peak Knee Flexion Ang Vel	Peak Hip Flexion Ang Vel	Max Angle Hip Flexion	Peak Knee Extension Ang Vel	Peak Hip Extension Ang Vel	Peak Hip Flexor Moment	Peak Hip Extensor Moment	Peak Knee Extensor Moment	Peak Knee Flexor Moment	Foot Velocity at Touchdown	Moment of Inertia of Leg	Stride Length	Stride Rate	100 m Time
Hip Angle at TO		-0.187	0.511	0.244	-0.224	0.277	0.494	0.128	-0.019	0.382	-0.423	0.372	-0.167	-0.323	0.219	0.521	-0.619	-0.257
Knee Angle at TO	-0.187		-0.765	-0.131	0.550	-0.100	-0.140	0.130	0.440	0.023	-0.347	0.163	-0.020	0.182	0.023	-0.053	-0.283	0.329
Max Angle Hip Extension	0.511	-0.765		0.513	-0.703	0.420	0.282	0.194	-0.354	0.217	-0.084	0.298	-0.004	-0.403	0.392	0.487	-0.454	-0.445
Minimum Knee Angle	0.244	-0.131	0.513		-0.364	0.578	0.064	0.581	0.233	0.279	-0.156	0.511	0.038	-0.308	0.863	0.487	-0.454	-0.496
Peak Knee Flexion Ang Vel	-0.224	0.550	-0.703	-0.364		0.063	-0.294	-0.014	0.082	-0.003	0.054	-0.090	0.100	0.004	-0.101	0.054	-0.499	0.544
Peak Hip Flexion Ang Vel	0.277	-0.100	0.420	0.578	0.063		-0.369	0.280	-0.328	0.408	-0.099	0.440	0.322	-0.177	0.734	0.054	-0.499	0.077
Max Angle Hip Flexion	0.494	-0.140	0.282	0.064	-0.294	-0.369		-0.006	0.238	-0.022	-0.169	0.074	-0.293	-0.591	-0.053	0.779	-0.330	-0.599
Peak Knee Extension Ang Vel	0.128	0.130	0.194	0.581	-0.014	0.280	-0.006		0.344	0.255	-0.337	0.357	0.199	-0.192	0.462	-0.030	-0.147	0.079
Peak Hip Extension Ang Vel	-0.019	0.440	-0.354	0.233	0.082	-0.328	0.238	0.344		0.043	-0.037	-0.290	-0.181	0.092	0.080	0.191	-0.021	-0.264
Peak Hip Flexor Moment	0.382	0.023	0.217	0.279	-0.003	0.408	-0.022	0.255	0.043		-0.144	0.085	0.142	0.225	0.141	0.099	-0.305	-0.052
Peak Hip Extensor Moment	-0.423	-0.347	-0.084	-0.156	0.054	-0.099	-0.169	-0.337	-0.037	-0.144		-0.490	0.196	0.202	-0.298	-0.284	0.625	-0.280
Peak Knee Extensor Moment	0.372	0.163	0.298	0.511	-0.090	0.440	0.074	0.357	-0.290	0.085	-0.490		-0.043	-0.383	0.549	0.382	-0.582	-0.114
Peak Knee Flexor Moment	-0.167	-0.020	-0.004	0.038	0.100	0.322	-0.293	0.199	-0.181	0.142	0.196	-0.043		-0.030	0.199	-0.307	0.085	0.416
Foot Velocity at Touchdown	-0.323	0.182	-0.403	-0.308	0.004	-0.177	-0.591	-0.192	0.092	0.225	0.202	-0.383	-0.030		-0.501	-0.555	0.420	0.274
Moment of Inertia of Leg	0.219	0.023	0.392	0.863	-0.101	0.734	-0.053	0.462	0.080	0.141	-0.298	0.549	0.199	-0.501		0.413	-0.610	-0.193
Stride Length	0.521	-0.053	0.366	0.487	-0.362	0.054	0.779	-0.030	0.191	0.099	-0.284	0.382	-0.307	-0.555	0.413		-0.698	-0.728
Stride Rate	-0.619	-0.283	-0.164	-0.454	-0.061	-0.499	-0.330	-0.147	-0.021	-0.305	0.625	-0.582	0.085	0.420	-0.610	-0.698		0.136
100 m Time	-0.257	0.329	-0.445	-0.496	0.544	0.077	-0.599	0.079	-0.264	-0.052	-0.280	-0.114	0.416	0.274	-0.193	-0.728	0.136	

3.3.4.1 Early Recovery

Leg recovery during sprinting begins with the slowing of the backwards rotation of the thigh and the rapid flexing of the knee to position the entire leg for forwards swing. Because of the angular momentum generated in the support leg while in contact with the ground, the thigh continues to rotate backwards slightly after takeoff. Figure 3-6 outlines the correlation between knee angle at takeoff and the maximum angle of hip extension. Sprinters who are more extended at the knee at takeoff tend to also extend more fully at the hip after takeoff. The rate of knee flexion during recovery has been found to be a predictor of sprinting speed in elite male and female sprinters (Alexander, 1989).

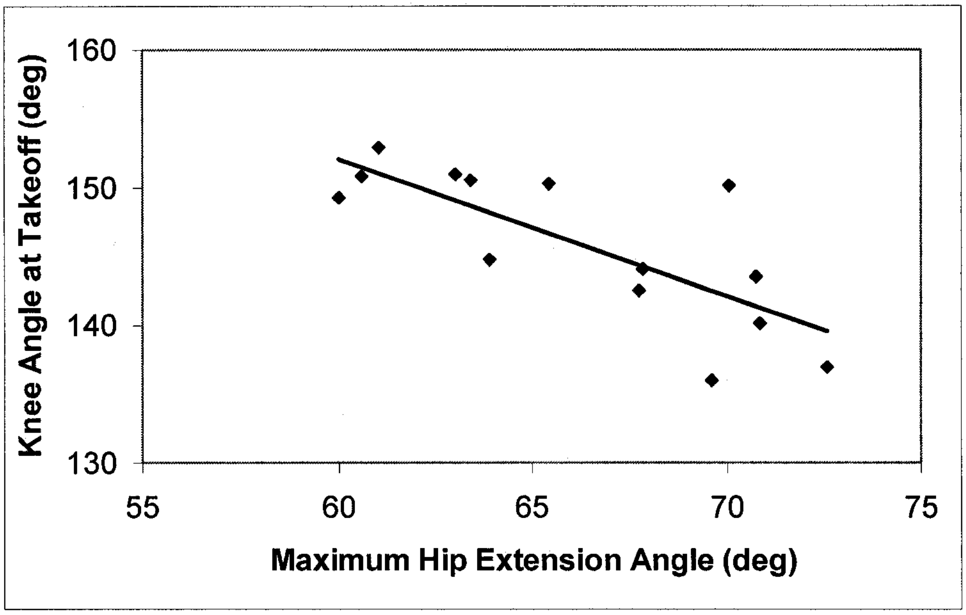


Figure 3-6. Correlation between knee angle at takeoff and maximum angle of hip extension ($r = -0.765$, $p < 0.01$).

Figure 3-7 shows the relationship between knee angle at takeoff and knee flexion angular velocity. Sprinters who minimize the knee angle at takeoff (smaller angle) do not flex the knee at as great a rate as those sprinters who are more extended at takeoff ($r = 0.550$, $p < 0.05$). Knee extension should be minimized at takeoff to reduce ground contact time (Mann, 1985), however, the knee angle at takeoff also determines how rapidly the knee is flexed into position for recovery. The greater the knee extension at takeoff the faster the knee must flex in order to achieve a proper position for forward swing. This action must be coupled with fast hip flexion angular velocity, which also influences the rate of knee flexion (Philips et al., 1983).

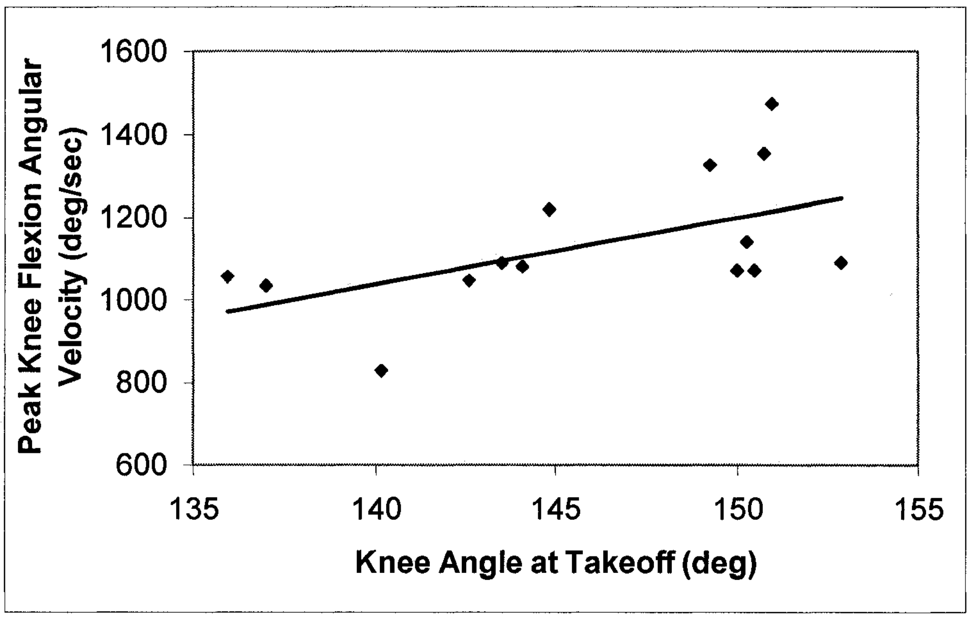


Figure 3-7. Correlation between peak knee flexion angular velocity and knee angle at takeoff ($r = 0.550$, $p < 0.05$).

A similar relationship was found between the maximum angle of hip extension and peak knee flexion angular velocity (Figure 3-8). The more extended the hip angle after takeoff, the faster the knee flexed in order to prepare the leg for forward swing ($r = -0.703$, $p < 0.01$). The association among these variables can be seen in the angle-angle diagram of the hip and knee for subjects 8 and 10 (Figure 3-9). At takeoff (Point 1), the two sprinters were in a

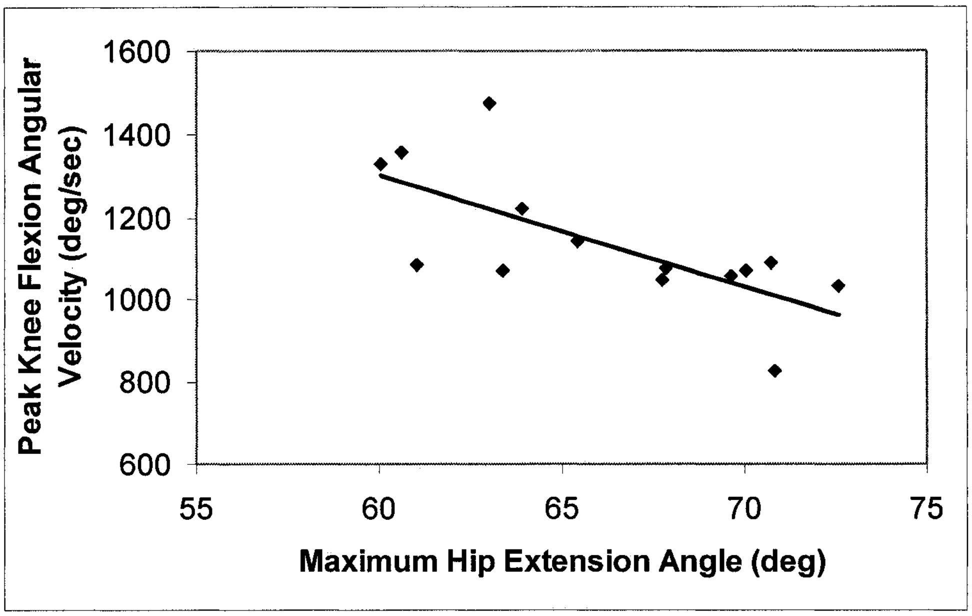


Figure 3-8. Correlation between peak knee flexion angular velocity and maximum angle of hip extension (r = -0.703, $p < 0.01$).

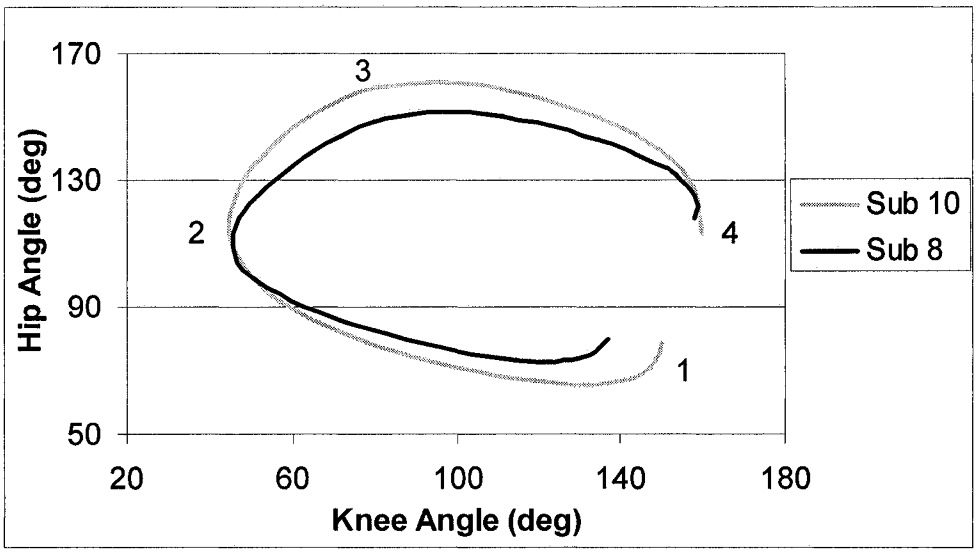


Figure 3-9. Angle angle diagram for the hip and knee for subjects 8 and 10. Point 1 = takeoff; Point 2 = minimum knee angle; Point 3 = maximum hip flexion; Point 4 = touchdown.

similar position at the hip but subject 10 was more extended at the knee. After takeoff, subject 10 showed a greater maximum angle of hip extension. Both athletes showed similar maximum angles of knee flexion (Point 2). For subject 10 to achieve the same angle of knee flexion as subject 8 despite being more extended at the knee at takeoff and at the hip after takeoff, it would be necessary for subject 10 to flex the knee faster. This was indeed the case, as subject 10 showed a peak knee flexion angular velocity of 1141°/s and subject 8 of 1030°/s.

3.3.4.2 Middle Recovery

Once the knee has fully flexed into the recovery position, which occurs when hip flexion angular velocity is maximal (Chengzhi & Zongcheng, 1987), it is the role of the knee extensor muscles to work eccentrically to control the knee angle as the thigh begins to swing forward (Dillman, 1971). The relationship between peak knee extensor moment and the minimum moment of inertia of the recovery leg is outlined in Figure 3-10. Sprinters with greater knee extensor moments also show smaller moment of inertia values, indicating that the more the knee is flexed, the more important the role of the knee extensors in controlling the movement. The function of the knee extensors during leg swing was also shown by Piazza and Delp (1996), in which they used computer simulation to investigate the influence of the leg musculature on the degree of knee flexion during the recovery phase of human gait. They found that increasing the knee extensor moment decreased the amount of knee flexion. Although the speed of locomotion and the amount of knee flexion were different between Piazza and Delp's study and the present one, there are similarities in the function of the knee extensor moment in controlling knee flexion.

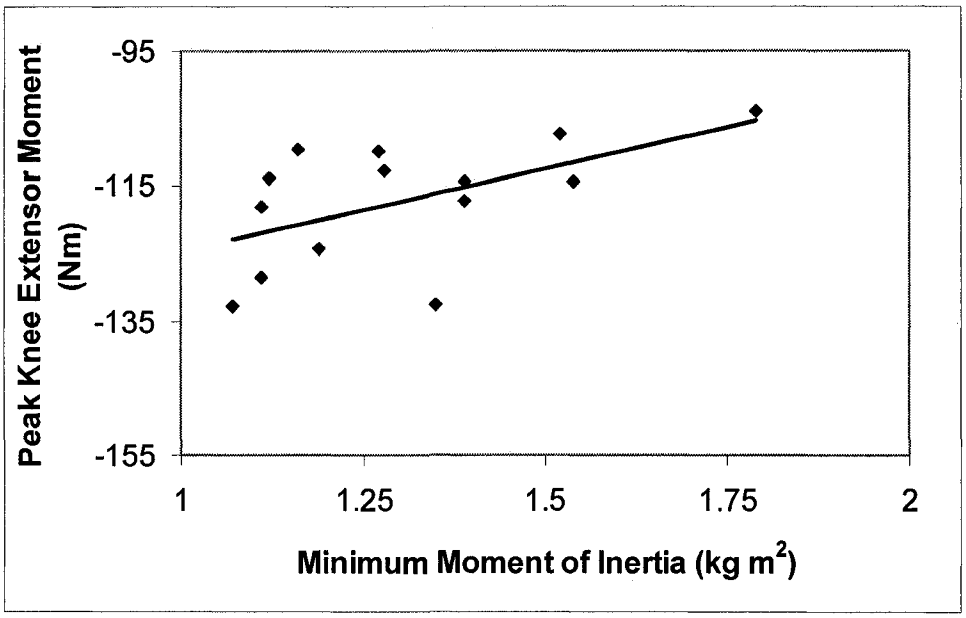


Figure 3-10. Correlation between peak knee extensor moment and minimum moment of inertia of the recovery leg ($r = 0.549$, $p < 0.05$).

Reducing the moment of inertia of the recovery leg about a transverse axis through the hip through greater knee flexion will decrease the resistance to rotation and enable the leg to swing through faster, which is supported by the results presented here. However, it is important to note what functions as the control mechanism for the moment of inertia of the recovery leg. The knee flexion movements in sprinting are not controlled by the hamstrings (Alexander, 1989). A major part of knee flexion occurs passively and results in a transfer of energy into the lower leg as a result of hip flexor activity (Chapman & Caldwell, 1983). Knee flexion during recovery is limited by the knee extensors, not initiated by the knee flexors (Mann, 1981). EMG analysis (Mero & Komi, 1987; Mann, Moran, & Dougherty, 1986; Simonsen, Thomsen, & Klausen, 1985) supports these statements in that no muscular activity of the hamstring muscles was found just after takeoff. Whether or not the quadriceps muscles function to control knee flexion is not clear. Studies from Mero and Komi (1987) and Simonsen et al. (1985) showed that the vastii muscles become active midway through recovery during the period of minimum knee angle, whereas Mann et al. (1986) indicated that activity in the quadriceps muscles is not seen until the latter stages of recovery. This means the control of knee flexion through the knee extensor moment could possibly be from two sources. The knee extensor moment could be a result of the muscle activity as suggested by Mero and Komi (1987) and Simonsen et al. (1985), it could result from the passive stretching of connective structures (tendons and ligaments) when the knee angle is minimized, or it could result from a combination of the two.

Knee flexion during recovery is one of the most recognizable and characteristic movements of the sprinting stride. Elite sprinters typically exhibit a fluid, cyclical leg action, and may or may not achieve a fully flexed knee position. The purpose of this movement is to reduce the moment of inertia of the recovery leg, thereby minimizing the resistance to rotation and allowing the leg to rotate forwards about the hip joint more quickly (Hay, 1993). The amount of knee flexion is closely related to the moment of inertia of the recovery leg - the smaller the minimum knee angle, the smaller the moment of inertia (Figure 3-11).

The importance of knee flexion in the recovery phase is outlined in Figures 3-12 and 3-13, which show the correlations between peak hip flexion angular velocity and the minimum moment of inertia of the recovery leg (Figure 3-12) and the minimum knee angle (Figure 3-13), respectively. The smaller the minimum

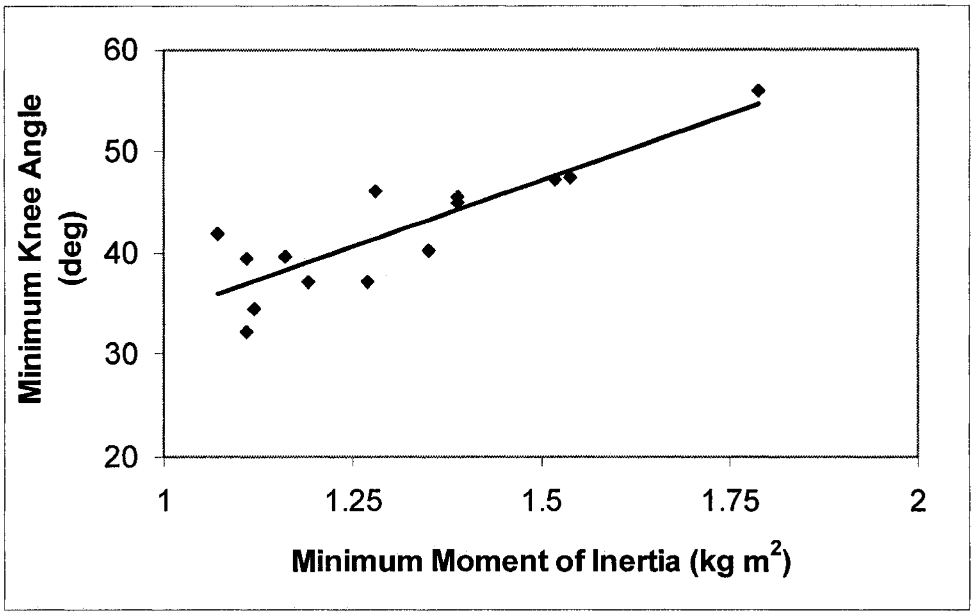


Figure 3-11. Correlation between minimum knee angle and minimum moment of inertia of the recovery leg ($r = 0.863$, $p < 0.01$).

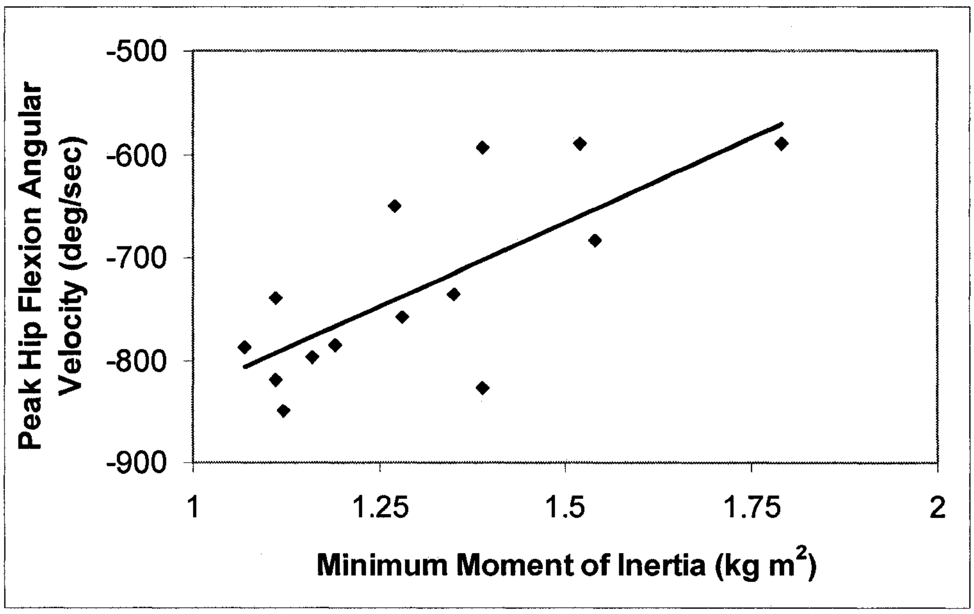


Figure 3-12. Correlation between peak hip flexion angular velocity and minimum moment of inertia of the recovery leg ($r = 0.734$, $p < 0.01$).

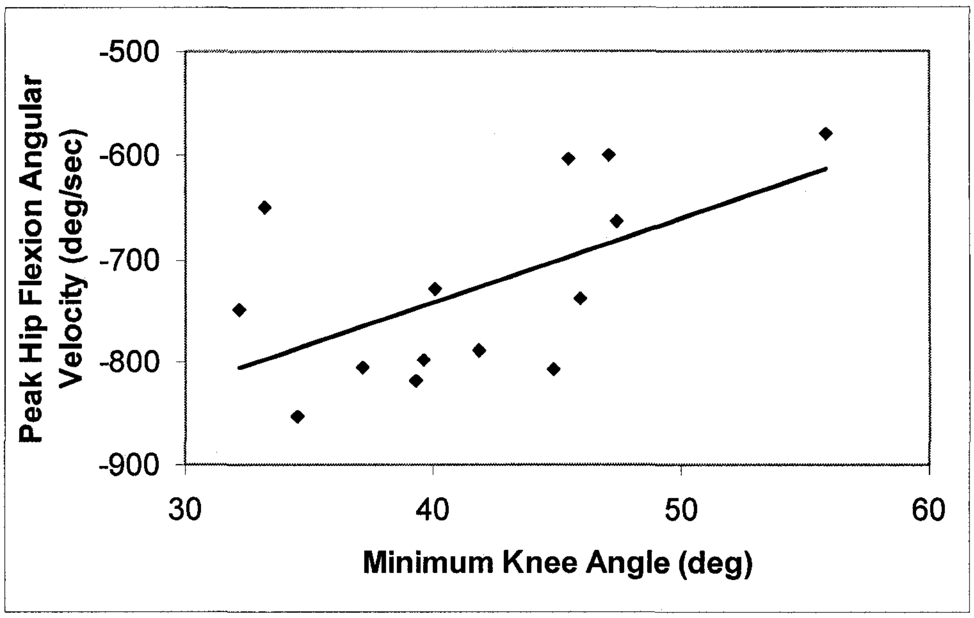


Figure 3-13. Correlation between peak hip flexion angular velocity and minimum knee angle ($r = 0.578$, $p < 0.05$).

knee angle and therefore the smaller the moment of inertia of the recovery leg, the faster the forward rotation of the leg about the hip. Sprinters who do not fully flex the knee experience a greater resistance to forward rotation and therefore do not recover the leg as rapidly.

One of the characteristics of less skilled sprinters is inadequate hip flexor strength, which results in a lack of full knee flexion often seen during recovery (Mann, 1985). From the results of the present study it is understandable how this conclusion could have been made, as sprinters who do not flex the knee fully also do not flex the hip as fast, and hip flexion angular velocity is a function of sprinting ability (Mann, 1985). A slower peak hip flexion angular velocity could be interpreted as being caused by a lack of power, when in fact it may be a function of both power and the athlete's sprinting mechanics. A larger rotational inertia would make it more difficult to swing the leg which could also result in a slower peak angular velocity.

When evaluating a sprinter's ability it is important to consider both strength and technique, as they mutually influence performance. The sprinters analyzed in the present study were some of the best in the world, and it can be assumed that they were very powerful athletes. The lack of full knee flexion observed in some of these individuals is not a product of inadequate strength, but a result of their sprint mechanics.

3.3.4.3 Late Recovery

As the hip approaches maximum flexion, the knee extends in order to prepare the foot for ground contact. Figure 3-14 outlines the correlation between knee extension angular velocity and minimum knee angle. Sprinters who flex the knee more during recovery also extend the knee at a greater rate ($r = 0.581$, $p < 0.05$). This increased knee extension angular velocity is a result of the larger knee extensor moment seen when the moment of inertia is smaller (refer to Figure 3-9). Sprinters who do not maximally flex the knee are in a position in which they do not need to rapidly extend the knee in order to prepare the foot for the subsequent ground contact.

The difficulty with this rapid knee extension arises when the lower leg must be slowed. The limiting factor in sprinting speed is seen when the sprinter is unable to develop sufficient eccentric muscle moment at the knee prior to ground contact. Consequently, a longer time must be spent in the air in order to properly prepare for contact, which changes the relationship between stride

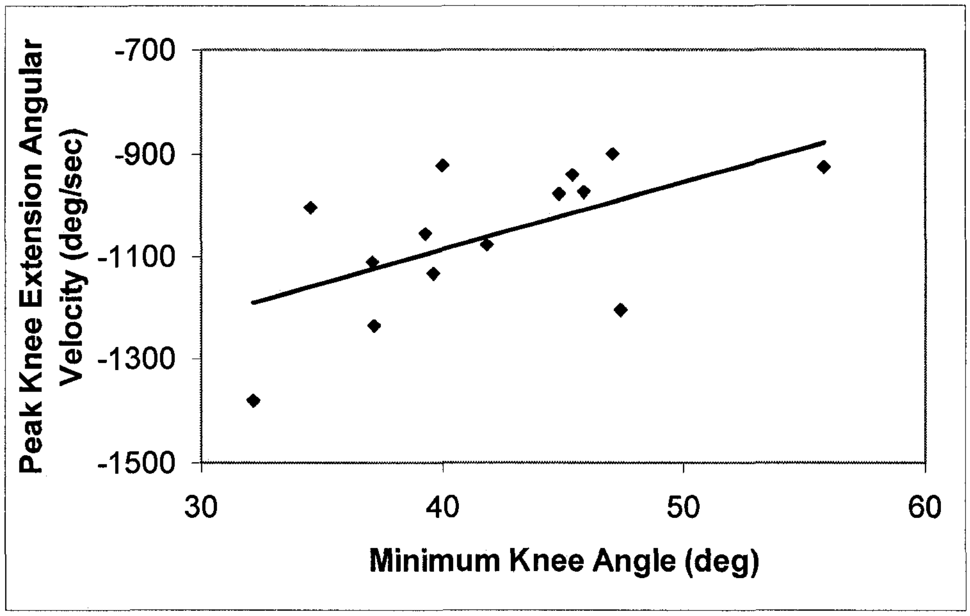


Figure 3-14. Correlation between peak knee extension angular velocity and minimum knee angle during recovery ($r = 0.581$, $p < 0.05$).

length and stride rate (Chapman & Caldwell, 1983). Even if the recovery leg was swinging in an optimal manner the eccentric contributions from the knee flexor muscles would be larger than physiologically possible (Wood et al., 1987). The results of this study show that one way that knee extension angular velocity could be reduced is by limiting the amount of knee flexion through recovery. This would decrease the eccentric knee muscle moment required to slow the knee extension, which in turn could reduce the limiting nature of this phase of recovery.

The horizontal velocity of the foot prior to ground contact determines if there is a braking effect at touchdown (Hay, 1993). The slower the horizontal foot velocity, the smaller the braking force on contact. A significant correlation was found between the horizontal velocity of the foot prior to touchdown and maximum hip flexion angle ($r = -0.591$, $p < 0.05$) (Figure 3-15), in that greater hip flexion angles are correlated with slower horizontal foot velocities. Anterior-posterior ground reaction forces often vary widely among sprinters (Wood, 1987), which is supported by the range of horizontal foot velocity values seen in elite sprinters (Kersting, 1999). These results show, however, that hip flexion is a necessary component of fast sprinting as it is related to the production of fast hip extension angular velocities and the reduction of braking on ground contact (Mann & Herman, 1985).

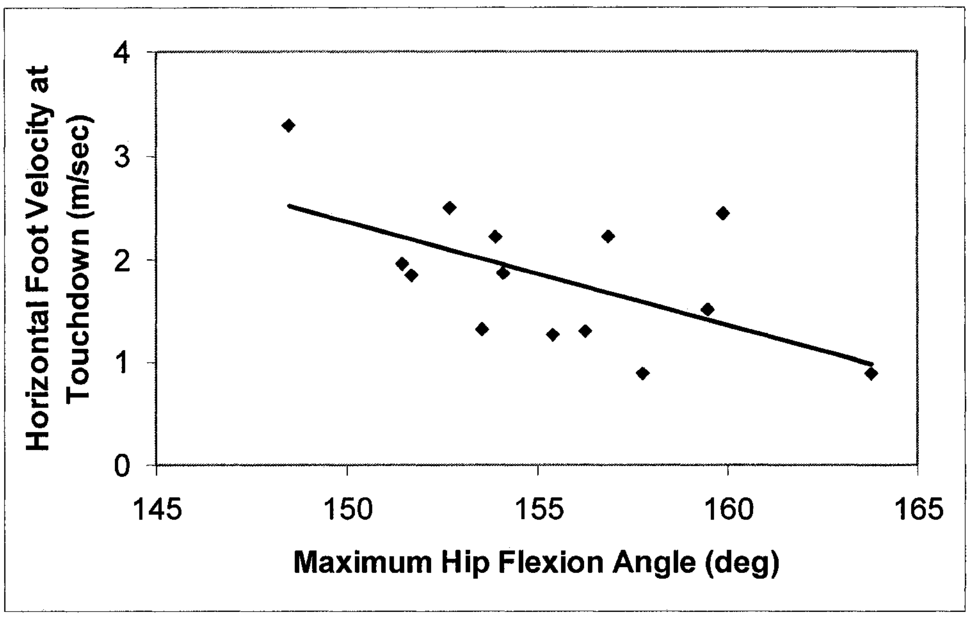


Figure 3-15. Correlation between horizontal foot velocity at touchdown and maximum hip flexion angle (r = -0.591, $p < 0.05$).

The relationship among the variables through the latter portion of the recovery phase can be explained by comparing the angle-angle diagrams of the hip and knee for subjects 7 and 10 (Figure 3-16). Because the knee was more flexed for subject 10 than for subject 7 during recovery (Point 2), it had to extend faster in order to move the foot forward in preparation for the subsequent ground contact. Peak knee extension angular velocity for subject 10 was 978°/s and for subject 7 was 926°/s. Also, because the knee was more flexed for subject 10, the recovery leg was able to swing through faster because of the decreased moment of inertia. This was seen in the faster peak hip flexion angular velocity for subject 10 of –827°/s compared to subject 7 of –589°/s.

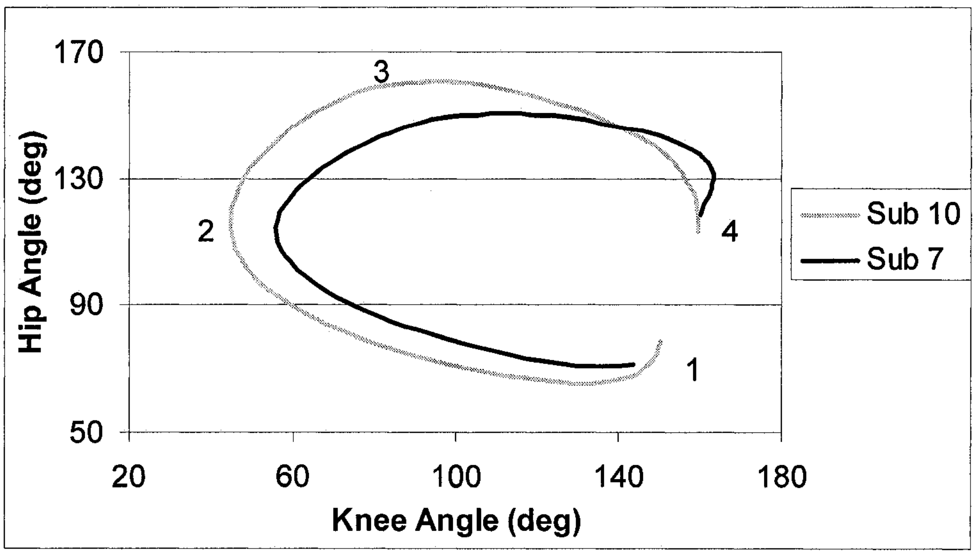


Figure 3-16. Angle-angle diagram of the hip and knee for subjects 7 and 10. Point 1 = takeoff; Point 2 = minimum knee angle; Point 3 = maximum hip flexion; Point 4 = touchdown.

The faster hip flexion angular velocity of subject 10 also allowed the hip to reach a more flexed position (Point 3). From this position, the sprinter has more time to prepare the foot for ground contact (Point 4), and a slower horizontal foot velocity prior to touchdown of 0.89 m/s was reported. In comparison, the slower hip flexion angular velocity of subject 7 did not allow the hip to flex the same extent, and the result was a slightly faster horizontal foot velocity of 1.39 m/s was found.

3.3.4.4 Stride Length and Stride Rate

Stride length and stride rate are fundamental aspects of human locomotion, and are mutually important factors to sprint performance. As described by Hay (1993), a number of individual variables interact to determine each individual's stride length and stride rate. Stride rate and stride length, in turn, function to determine the average sprinting velocity. A number of statistically significant relationships were found in this study between both stride length and stride rate and other variables related to sprinting performance.

The relationship between stride length and stride rate has been described previously (Luhtanen & Komi, 1978; Hunter et al., 2004), in which faster stride rates are associated with shorter stride lengths, and vice versa. A similar relationship was found in this study ($r = -0.698$, $p < 0.01$). At the point of takeoff, a significant relationship was found between hip angle at takeoff and stride rate, in which the more extended the hip angle at takeoff, the slower the stride rate ($r = -0.610$, $p < 0.05$). The more extended the hip is at takeoff, the more time it takes to swing the leg forwards because the hip must rotate through a greater range of motion, resulting in a slower stride rate.

Through the middle portion of recovery, significant relationships were found between stride rate and both the minimum moment of inertia of the recovery leg and the peak knee extensor moment. The smaller the moment of inertia of the recovery leg, the faster the stride rate ($r = -0.610$, $p < 0.05$). Also, the larger the peak knee extensor moment, the faster the stride rate ($r = -0.582$, $p < 0.05$). These relationships are associated with the correlations discussed previously among the peak knee extensor moment, minimum moment of inertia of the recovery leg, and hip flexion angular velocity. Larger knee extensor moments are seen when the moment of inertia of the recovery leg is minimized. The moment of inertia of the recovery leg is also correlated with the peak hip flexion angular velocity. These variables function together to increase the stride rate.

Late in the recovery phase, the hip reaches maximum flexion and begins to extend to bring the foot towards the ground for contact. This action is controlled by the extensor moment acting at the hip. Peak hip extensor moment was found to be significantly correlated to stride rate, in which larger extensor moments were seen during higher stride rates ($r = 0.625$, $p < 0.05$). Peak extensor moment was not found, however, to be correlated with peak hip extension angular velocity, so it cannot be said that a larger extensor moment will result in a faster hip extension and a higher stride rate. It is possible that the relationship between peak hip extensor moment and stride rate is seen in the time it takes to slow the forward rotation of the thigh at the hip and initiate hip extension. Larger hip moments may act to increase the rate of hip deceleration/acceleration which would increase the stride rate.

As the hip reaches a maximum angle of flexion, a significant relationship was found between maximum hip flexion angle and stride length. The greater the maximum hip flexion angle, the longer the stride length ($r = 0.779$, $p < 0.01$). Similarly, stride length was found to be related to the horizontal velocity of the foot immediately prior to ground contact, in which longer stride lengths were correlated with slower horizontal foot velocities ($r = -0.555$, $p < 0.05$). These results are similar to those discussed earlier between maximum hip flexion angle and horizontal foot velocity, and it is evident that there is a relationship among these three variables. An increased hip flexion angle would allow the sprinter the time to position foot prior to contact, which would minimize the horizontal foot velocity and result in a longer stride length. These correlations support the findings of others who have stated that this portion of the recovery phase is important to sprint performance (Chapman & Caldwell, 1983; Wood, 1987).

The correlation coefficient is designed to measure linear relationships between two variables on an interval scale, and it is assumed that the data is normally distributed. This study showed that there are a number of important relationships among the kinematic and kinetic variables which govern the movement of the recovery leg during sprinting, as was seen by statistically significant linear correlations which ranged from ± 0.549 to ± 0.863. Similar correlation values have been found in previous sprint studies involving comparable numbers of subjects (Nummela, Rusko, & Mero, 1994; Alexander, 1989; Mero & Komi, 1986).

It is important to note that correlations are often misleading in suggesting a stronger relationship than actually exists. When interpreting the results, both the correlation coefficient (r) and the coefficient of determination (r^2) should be considered. The coefficient of determination represents the proportion of common variation shared between two variables (Hassard, 1991), or more simply, the "strength" or "magnitude" of the relationship. In this study the percentage of the variance shared was between 30.1% and 74.5%, which are relatively low values. Rummel (2006) stated that when interpreting the correlation coefficient, it is important to know what one is testing or concerned about. In order to make predictions from the results, high correlations are necessary in order to minimize the unwanted variance. If one is concerned about uncovering relationships, however, then statistical significance is adequate. This study was designed to simply determine the relationships among the variables associated with the recovery phase of sprinting. Therefore, it is believed that the statistically significant correlations have been interpreted appropriately. The sample size was relatively small in this investigation. Ideally, more subjects would have been included in the analysis in order to increase the statistical power. This study was focused on a very specific population during competition, which made it virtually impossible for the investigator to control the testing environment and the number of subjects which could be included in the analysis.

When studying athletes, there are various factors which must be taken into account when considering how the results can be applied. Because the video for this study was collected during these two high level competitions, it was expected that the athletes were performing at their maximal effort. If the video were collected during a training session or a less significant competition, it would be likely that the performances would not be at the same level of intensity and performance. In addition, this study focused on the maximum velocity section of the race, which occurs between 40 and 80 metres (Ae et al., 1992). This is one of phases describing the 100 metre sprint race, which may be divided into the acceleration, maximum velocity, and deceleration phases (Schmolinski, 1996). Statistically significant correlations were found between 100 metre time and peak knee flexion angular velocity ($r = 0.544$, $p < 0.05$), maximum angle of hip flexion ($r = -0.599$, $p < 0.05$), and stride length ($r = -0.727$, $p < 0.01$). These correlations suggest that these variables may be important in running fast sprint times, however, because the data was collected only through the maximum velocity portion of the race it is difficult to draw conclusions on the overall performance outcome based on the kinematics from one phase. The results of this study are only applicable to the maximum speed section of the race. No research has examined the lower extremity kinematics and kinetics throughout the different phases of the 100 metre sprint race, but it is likely that there would be differences seen. The time of year and the phase of the athletes' training cycle would also have an influence on the results of the analysis. Sprint performance varies throughout the season as the athletes attempt to achieve a peak performance at the time when it means the most, such as during a national or world championship. For example, Table 3-8 shows the 100 metre results for one of the participants in this study from various competitions in the months leading up to the World Championships in Athletics. This athlete achieved one performance peak at the United States Track and Field Championships, and a second at the World Championships.

The results of this study have a direct application to understanding the relationships among the kinematic and kinetic variables which control the movements of the recovery leg in elite male sprinters. It is unclear if these

Table 3-8. 100 metre race results from one participant in the months prior to the World Championships.

Date	100 metre time
May 12	10.17
June 11	10.07
June 22	9.96
June 29	10.21
July 6	10.15
July 22	10.13
August 5	9.94

relationships would be the same if examining other populations such as non-elite level sprinters because of differences in the kinematics (Kivi, 2002b; Mann, 1985) and kinetics (Vardaxis & Hoshizaki, 1989) among individuals of different abilities. If technique differs among elite sprinters, however, it is likely that variations would be seen among athletes of different abilities as well. If this is the case, it is plausible that there would be relationships among the kinematic and kinetic variables among non-elite sprinters, and that some of these relationships would be similar to those reported in this study. At this time, this question remains unanswered and warrants further investigation.

3.4 Conclusions and Implications for Training

It has been suggested that the ultimate speed limiting factor in sprinting is leg recovery (Wood, 1987; Chapman & Caldwell, 1983). However, Mann (1986) said that there is considerable variation in the action of the lower leg during recovery. From the results of this investigation, it is evident that despite this variability there are definable movement patterns and relationships among performance variables seen in the recovery phase of sprinting. It was hypothesized that takeoff position at the hip and knee would influence the degree of knee flexion. It was found that both the amount of knee extension at takeoff and the maximum hip extension angle are correlated with the rate of knee flexion. Greater knee flexion angles are correlated with faster hip flexion angular velocities, which was an anticipated finding. However, knee extensor moments and knee extension angular velocities were found to be related to hip flexion angular velocity as well. Peak hip flexion angle was found to be correlated to both the horizontal velocity of the foot immediately prior to touchdown and stride length.

Because of the number of relationships found among the performance variables associated with the recovery phase of sprinting, and the fact that the relationships were seen throughout recovery, it is possible that changes in one aspect of the leg movement may result in changes to another. Coaches and athletes must be aware of this fact to understand both the positive and negative consequences of changing technique. It must be noted, however, that the correlations found in this analysis do not indicate that modifications in one aspect of the movement will cause changes somewhere else. The results simply state that these sprint-specific variables are related to each other.

When working with elite athletes, whether it is as a coach or scientist, it is often believed that the athlete is performing the movement or skill "the best way." Previous sprint biomechanics studies have identified the characteristics of elite sprinters that are most often associated with high-level performance (i.e., Mann, 1985; Mann & Herman, 1985). The results of this analysis showed that there may not be one "best" way of executing a movement such as swinging the leg forward during sprinting. It is up to the coach and scientist to understand the strengths and weaknesses of each individual athlete, and to know that it is not always necessary to perform a skill in a way that is theoretically ideal.

This study was conducted in order to develop a better understanding of individual differences among elite sprinters. This research, however, has application in the analysis of other movements and in the understanding of individual differences among athletes in other sports as well. The term "range of correctness" is often used when observing motion (Knudson & Morrison, 1997), referring to the fact that there are individual differences in the technique used by athletes which must be considered when analyzing a movement. This study showed that there was a range of correctness in the kinematics and kinetics of the recovery leg of elite sprinters, and despite these differences were a number of significant correlations among the variables. This concept can be expanded and used to understand individual differences among athletes performing other movements as well. This type of analysis would allow for the quantification of the relationships among the kinematic and kinetic variables governing a specific movement, which would be important when considering the individual differences typically seen in the performance.

This study has established the groundwork for future research into the investigation of individual differences in the biomechanics of sprinting. Having quantified the relationships among the kinematic and kinetic variables which govern motion through the flight phase, it is necessary to expand this research to include a similar analysis during the contact phase. This will involve examining the relationships among the lower extremity kinematics and kinetics during the contact phase in order to determine how variations in technique during the flight phase influence the contact phase, and to investigate how individual differences during both the flight and contact phases affect the production of ground reaction forces.

CHAPTER 4

Project 2 – Simulation of the Recovery Leg During Sprinting

4.1 Purpose

The purpose of this project was to investigate how changing the takeoff position of the recovery leg during sprinting would influence the subsequent lower extremity kinematics. In order to perform this analysis a simulation model had to be developed. It was hypothesized that more hip and knee extension at takeoff would lead to small knee flexion angles because the rotational inertia of the leg would need to be reduced to allow the leg to recover adequately. This reduced moment of inertia would result in faster hip flexion angular velocities and larger angles of hip flexion. Conversely, less extension at the hip and knee at takeoff would result in less knee flexion. This would increase the rotational inertia and result in slower hip flexion angular velocities and smaller angles of hip flexion.

4.2 Methodology

4.2 1 Development of Simulation Model

The computer simulation model used in this study was developed using the Working Model® 2D 5.03 (Knowledge Revolution, San Mateo, CA) software package. A subject-specific, planar, three segment model comprising of a thigh, shank, and foot segment was used to simulate the leg during the recovery phase of sprinting for all 14 subjects and was based on the data as calculated from Project 1. Body segment parameter data from Zatsiorski and Seluyanov (1983) as modified by DeLeva (1996) was used to estimate segmental masses, centre of mass locations (measured from proximal end of a segment), and moments of inertia. Segmental lengths were determined from the kinematic data. Initial joint angle, angular velocity, horizontal velocity of centre of mass, and vertical velocity of centre of mass for each segment were taken at the point of takeoff from the kinematic data and used as initial conditions in the simulation model.

In order to more accurately simulate the sprinting motion, a fourth segment was added which consisted of the remaining mass of the body and was represented by a solid mass. The entire system moved along a slot joint on a frictionless pin, with the slot following a parabolic path which was determined from the horizontal and vertical displacement of the hip during sprinting. The body segment moved with the horizontal velocity that was measured from the kinematic data at the hip joint. A screen capture of the model is seen in Figure 4-1. The system was constrained to moving along the slot joint because the simulation model was comprised of only one leg and there was no contralateral leg to prevent the system from falling due to gravity. The simulation was of the recovery phase of one complete sprint stride, which consists of two steps.

Joints were represented by frictionless pins, with torque generators located at the joints to produce movement. This is the same method used by Philips et al. (1983) and Wood et al. (1987) in their simulations of the recovery phase of sprinting. A Newtonian approach was used to derive the equations of motion, which states the sum of the forces acting are equal to the mass times the acceleration for linear movement, and the sum of the moments acting on a segment are equal to the moment of inertia times the angular acceleration for

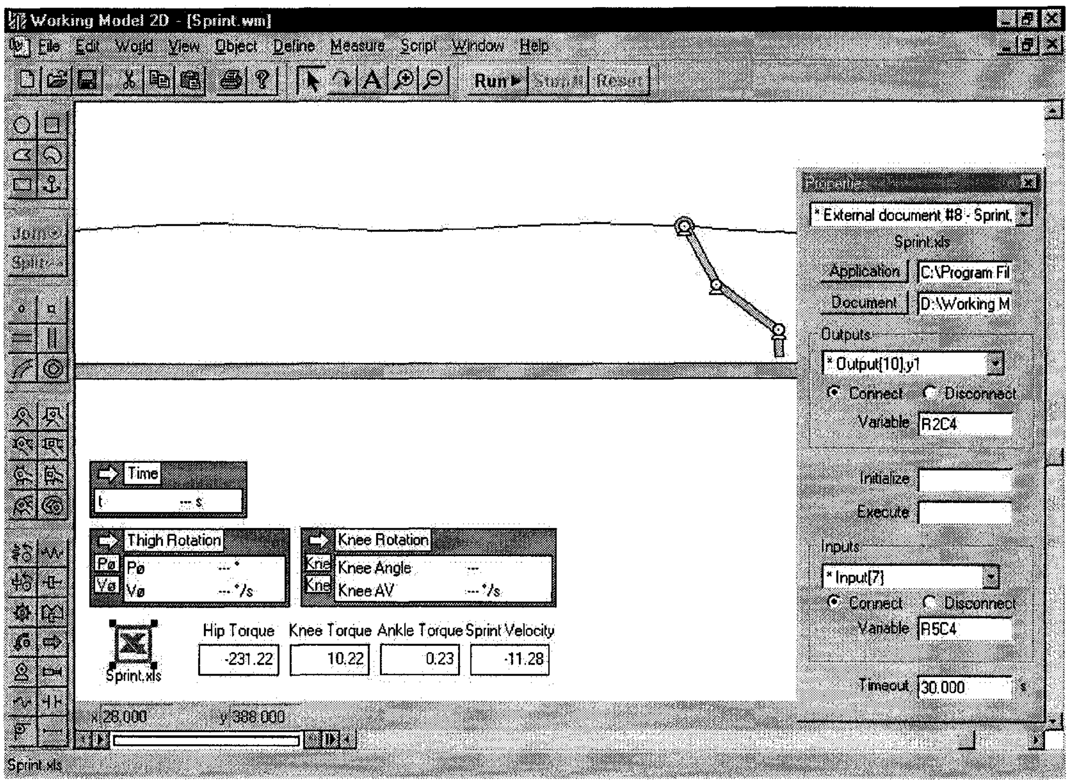


Figure 4-1. Screen capture of simulation model developed using Working Model 2D.

rotational movement. Refer to Equations 1 to 9 in the Methodology section of Project 1 for a complete description. All the input values to the model, including the torque values and initial horizontal and vertical velocities of the 4 segments, were put into a Microsoft Excel file. The input values were read via Dynamic Data Exchange (DDE) from the spreadsheet to run the simulation. The torque and velocity data were input into the simulation as discretized values at intervals of 0.00833 seconds. The trajectories of the three leg segments were not constrained in any way. Segments were free to rotate beyond anatomical limits in order to provide information as to whether or not specific hip and knee takeoff angles would be injurious to the athlete through recovery. Numerical integration was completed using the Runge-Kutta algorithm.

4.2.2 Validation of Working Model 2D as a Simulation Program

This software is a popular modeling and simulation program and has been used previously in studies involving gait and locomotion (Yakovenko, Gritsenko, & Prochazka, 2004; Ferris, Liang, & Farley, 1999). In order to validate the use of Working Model 2D as a computer simulation program, four tests were completed. The first was a swinging pendulum test, in which the period of one oscillation was calculated. The second test was a ball drop test to determine the vertical velocity immediately prior to contact with the ground from a known height. The third test examined the torque applied to a body and its resulting motion. The fourth test was a double pendulum in which the angular displacement was measured. These tests were selected because they are all based on fundamental laws of physics, and the theoretical value could be determined and compared to the simulation results.

Test 1 – Swinging Pendulum

The period of oscillation for a pendulum is determined by

$$T = 2\pi\sqrt{\frac{L}{g}}$$

where

T is the period of oscillation
L is the length of the pendulum
g is the acceleration due to gravity

For pendulum lengths of 0.5m, 1.0m, and 1.5m, this would result in oscillation periods of 1.42 sec, 2.01 sec, and 2.46 sec, respectively, for amplitudes of less than 22° (Lewowski & Wozniak, 2002). When pendulums of the same length were simulated using Working Model 2D, the oscillation periods were replicated exactly for the three conditions.

Test 2 – Ball Drop

For an object being dropped from a known height (d) with an initial velocity of 0 m/s, it's velocity at contact with the ground will be

$V_f^2 = v_i^2 + 2ad$

where

v_f is the final velocity
v_i is the initial velocity
d is the height above the ground
a is the acceleration due to gravity (-9.81 m/s^2)

If a ball is dropped from a height of 2.5 metres, it will contact the ground with a vertical velocity of –6.26 m/s. Similarly, a ball dropped from a height of 5.0 metres will contact the ground with a velocity of –9.90 m/s. When these two conditions were simulated using Working Model 2D, these theoretical values were reproduced exactly.

Test 3 – Torque and Motion Relationship

From the angular equivalent of Newton's Second Law of motion, it is known that

$T = I\alpha$

and can be rewritten as

$$\alpha = \frac{T}{I}$$

which states that the angular acceleration of a body (α) is directly proportional to the net torque (T) acting on the body, in the direction the torque is applied, and is inversely proportional to the moment of inertia (I) (Hall, 1995). A simulation model was developed in which a torque of 5 Nm was acting on a body rotating about it's centre of mass, and the body had a moment of inertia of 0.5 kgm^2. When the simulation was run, the body rotated with an angular acceleration of 10 m/s^2.

Test 4 – Double Pendulum

A double pendulum consists of one pendulum attached to the end of another. A simulation of a double pendulum was developed using Working Model 2D, and a program describing the motion of a double pendulum using Matlab was obtained from the internet (http://homepages.cae.wisc.edu/~cs310/matlab/). For both simulation methods, the masses of the two pendulums were each 0.5 kg, and were separated by rigid massless rods 0.5 m in length. The top pendulum (Pendulum 1) was positioned at an angle of 60° from the vertical, and the bottom pendulum (Pendulum 2) was positioned at an angle of 45° from the vertical. Comparisons were made between the angular displacements of the two pendulums for the two simulation methods for one cycle. The results showed that Working Model 2D replicated the motion of the double pendulum as presented by the Matlab program (Table 4-1)

Table 4-1. Comparison of angular displacement data for a double pendulum simulation using Working Model 2D and Matlab;

	Working Model 2D	**Matlab**
Pendulum 1	109.8°	110.2°
Pendulum 2	114.4°	115.7°

These tests, although simple in design, were all based on fundamental laws of motion and had results which were easily predicted which made them ideal for validating Working Model 2D as a simulation program. Working Model

2D accurately replicated the results of each test, indicating that it is a valid program for simulating movement.

4.2.3 Model Validation

To determine the validity of the simulation model, comparisons were made to gait data from a known source and to the kinematic data from Project 1. To confirm the efficacy of the modeling procedure, a model was set up using data presented by Winter (1979). Anthropometric data (segment lengths, masses, and moments of inertia) was used to construct the four link segments, initial conditions (joint angles, joint angular velocities, horizontal and vertical linear velocities) were applied, and joint moments were input to run the simulation for the swing phase of gait. The resulting angular displacement output data was comparable to that reported by Winter (1979), with small deviations seen near the end of the swing phase (Figures 4-2 and 4-3). Statistical analysis was

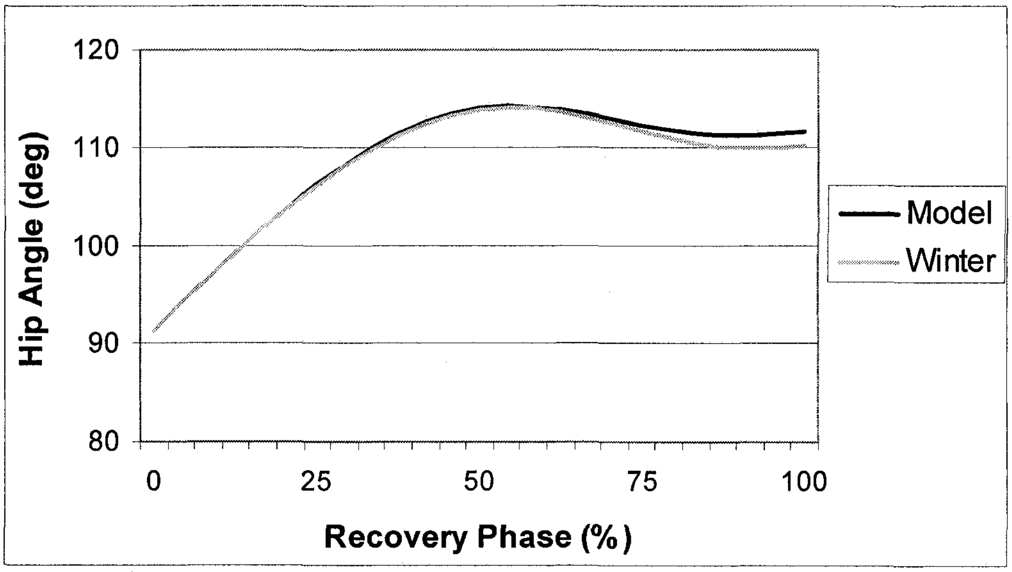


Figure 4-2. Hip angle comparison between simulation and data from Winter (1979).

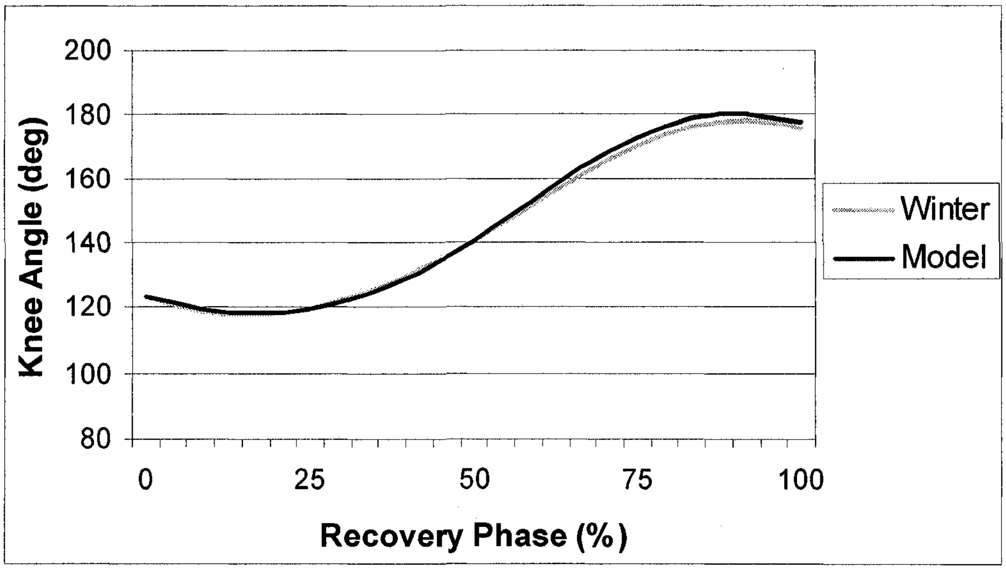


Figure 4-3. Knee angle comparison between simulation and data from Winter (1979).

completed using SPSS (Version 13). Intraclass correlation coefficients showed a high degree of reliability between the model and Winter's data for both the hip (ICC = 0.88) and knee (ICC = 0.98).

The second method of validating the model was by comparing the output data from the simulation to that of the kinematic data from Project 1. Figures 4-4 and 4-5 show the comparisons between the mean hip and knee angle values through the recovery phase from the video analysis and the simulation model for all subjects. As was the case for the angular velocities, small differences were seen after takeoff and before contact. Figures 4-6 and 4-7 show the mean hip and knee angular velocity values through the recovery phase from the video analysis and the simulation model for all subjects. The model closely replicated the kinematic of the sprinter, with small deviations seen in the peak angular velocity values and prior to ground contact. Intraclass correlation coefficients were calculated to determine the reliability of the experimental and the simulation data for the angular velocities and angular displacements at the hip and knee. All variables showed ICC values greater than 0.98.

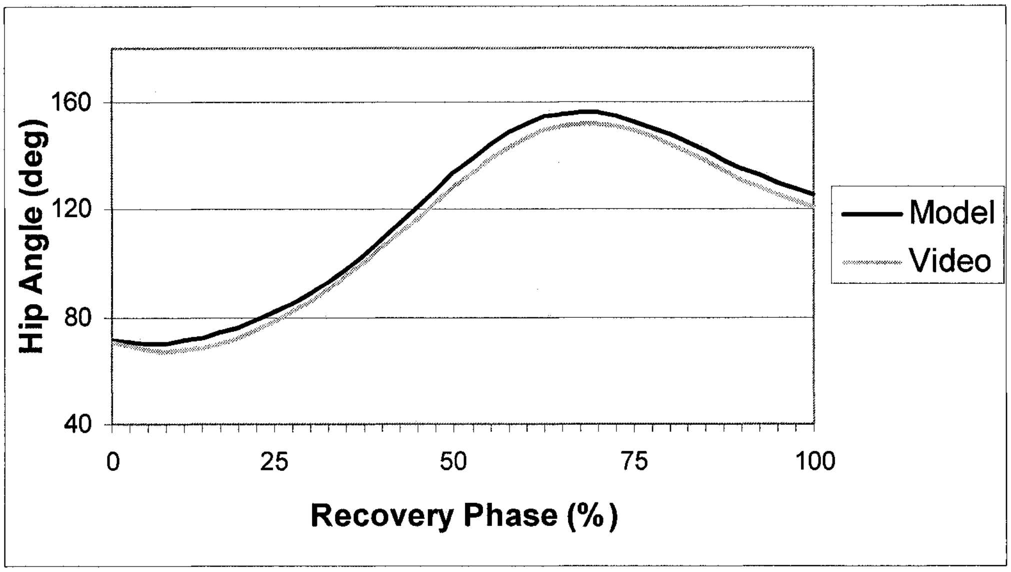


Figure 4-4. Mean hip angle comparison between video and simulation model for all subjects.

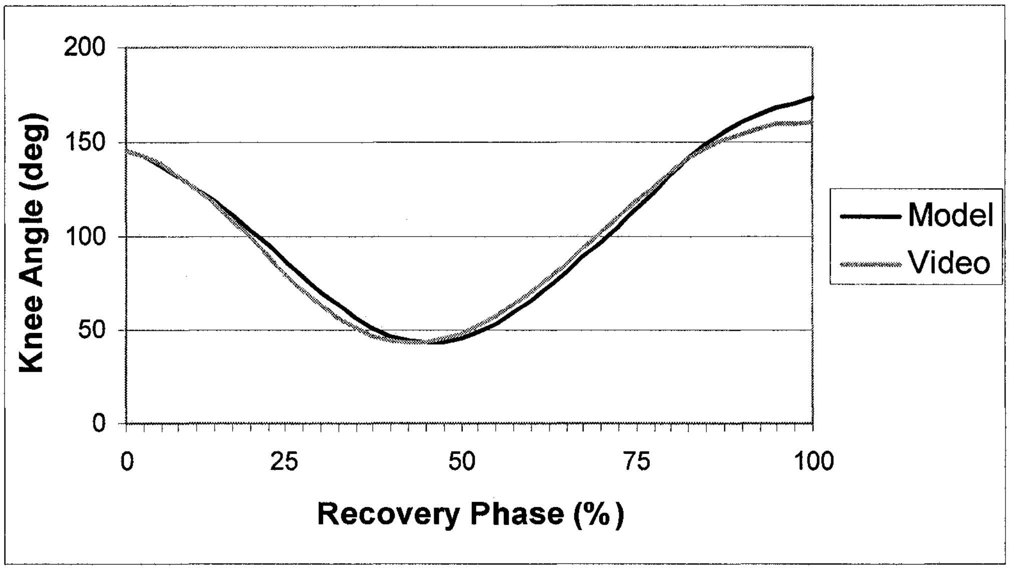


Figure 4-5. Mean knee angle comparison between video and simulation model for all subjects.

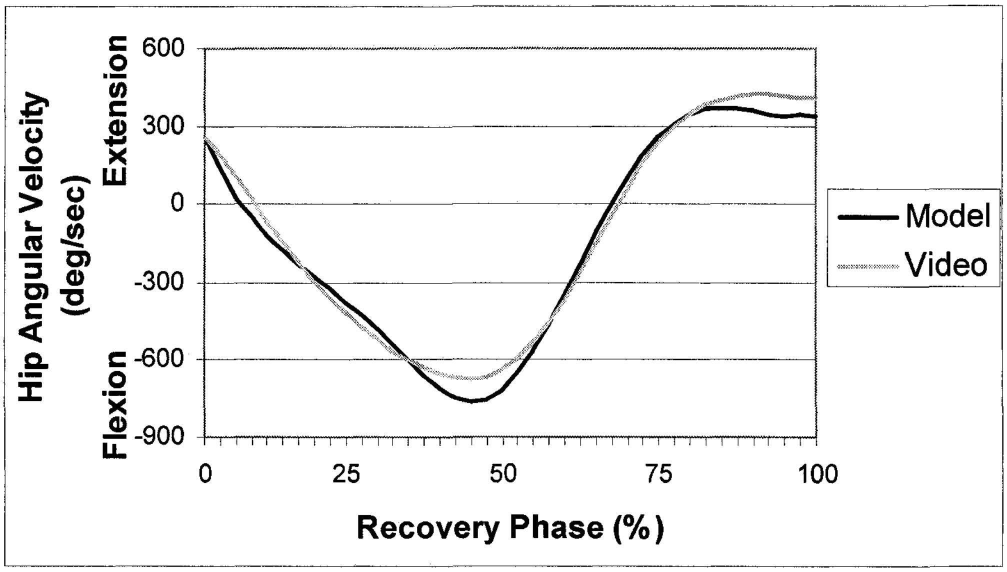


Figure 4-6. Mean hip angular velocity comparison between video and simulation model for all subjects.

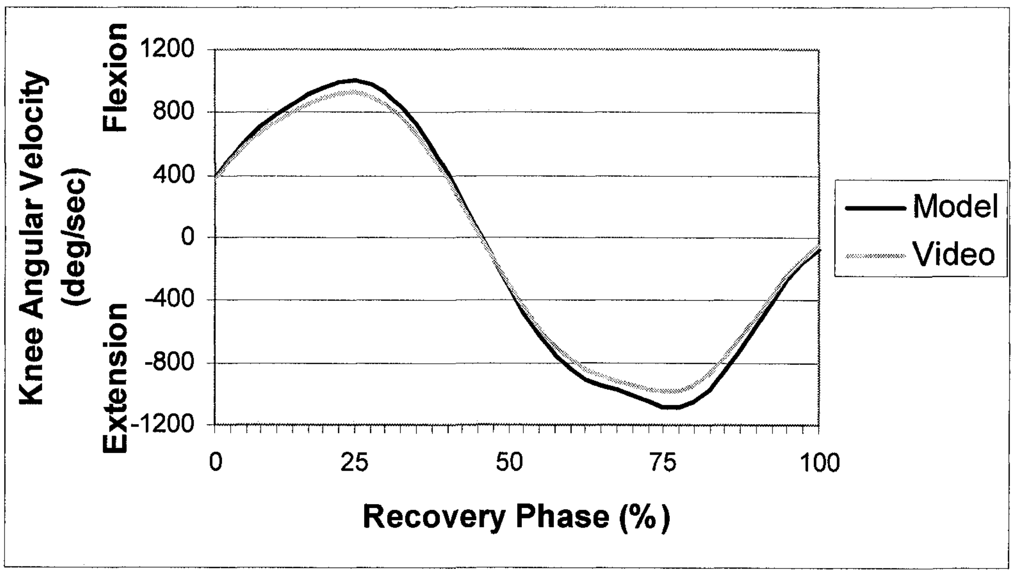


Figure 4-7. Mean knee angular velocity comparison between video and simulation model for all subjects.

Table 4-1 presents segmental length (L), mass (m), and moment of inertia (I) data used to develop the model for each subject. The participants in this study were a homogeneous group of elite sprinters; accordingly, the anthropometric data were similar across participants. This data, used in combination with the kinematic and kinetic inputs (Appendix 1), is a valid method of simulating the movements of the recovery leg of elite sprinters. Moreover, the strong intraclass correlations found between the Winter gait data and the simulation used for validation suggests that this simulation method can be used for the construction of a general, two-dimensional model of the lower extremity during the swing phase of gait.

Table 4-2. Anthropometric data used for each individual simulation model.

Subject	L Thigh (m)	L Shank (m)	L Foot (m)	m Thigh (kg)	m Shank (kg)	m Foot (kg)	I Thigh (kgm^2)	I Shank (kgm^2)	I Foot (kgm^2)
1	0.46	0.52	0.21	10.34	3.16	1.00	0.24	0.05	0.003
2	0.44	0.45	0.23	11.33	3.46	1.10	0.24	0.04	0.003
3	0.43	0.45	0.21	10.62	3.25	1.03	0.21	0.04	0.003
4	0.42	0.44	0.21	10.62	3.25	1.03	0.20	0.04	0.003
5	0.44	0.45	0.20	10.90	3.33	1.05	0.23	0.04	0.003
6	0.44	0.51	0.21	10.62	3.25	1.03	0.22	0.05	0.003
7	0.45	0.50	0.23	11.75	3.59	1.14	0.26	0.06	0.004
8	0.43	0.46	0.22	11.75	3.59	1.14	0.24	0.05	0.003
9	0.44	0.47	0.21	9.91	3.03	0.96	0.21	0.04	0.003
10	0.45	0.45	0.22	11.75	3.59	1.14	0.26	0.05	0.003
11	0.44	0.46	0.20	11.33	3.46	1.10	0.24	0.05	0.003
12	0.44	0.47	0.21	9.77	2.99	0.95	0.20	0.04	0.003
13	0.43	0.47	0.21	10.62	3.25	1.03	0.21	0.04	0.003
14	0.44	0.46	0.23	10.62	3.25	1.03	0.22	0.04	0.003

4.2.4 Simulation of Recovery Leg

In order to determine what effects changing the hip and knee angles at takeoff would have on the subsequent leg kinematics for one sprinter, the initial conditions were adjusted as follows:

Hip 0	Knee 0
Hip+3	Knee+3
Hip+6	Knee+6
Hip-3	Knee-3
Hip-6	Knee-6

The Hip 0 and Knee 0 conditions were those determined for each subject from Project 1 and were used as initial conditions during the initial simulation. The Hip+3 and Hip+6 conditions were those in which the hip was in a position of greater extension by 3° and 6° respectively (i.e., smaller numerical values representing the absolute joint angles as measured from the right horizontal) as compared to the initial sprinting condition, and the Hip-3 and Hip-6 conditions were those in which the hip was more flexed by 3° and 6° (i.e., larger numerical values representing the absolute joint angles as measured from the right horizontal). The same adjustments apply to the knee, with the positive (Knee+3 and Knee+6) values representing greater knee extension by 3° and 6°, and the negative values (Knee-3 and Knee-6) representing less knee extension at takeoff. The maximum change to hip and knee angles was ±6° from the initial condition. Mann (1985) used an angle of approximately 6° to differentiate among "good," "average," and "poor" sprinters in hip and knee angles at takeoff. This maximum value was also selected to keep the model within realistic angles seen during sprinting. Hip, knee and ankle torque was not modified in any way during the simulation.

The computer simulation was then run in the following combinations of initial hip and knee angles:

1. Hip 0, Knee 0 (Initial condition)
2. Hip+3, Knee 0
3. Hip+6, Knee 0
4. Hip-3, Knee 0
5. Hip-6, Knee0
6. Hip 0, Knee+3
7. Hip 0, Knee+6
8. Hip 0, Knee-3
14. Hip+6, Knee+3
15. Hip+6, Knee+6
16. Hip+6, Knee-3
17. Hip+6, Knee-6
18. Hip-3, Knee+3
18. Hip-3, Knee+6
20. Hip-3, Knee-3

9. Hip 0, Knee-6	21. Hip-3, Knee-6
10. Hip+3, Knee+3	22. Hip-6, Knee+3
11. Hip+3, Knee+6	23. Hip-6, Knee+6
12. Hip+3, Knee-3	24. Hip-6, Knee-3
13. Hip+3, Knee-6	25. Hip-6, Knee-6

In order to more accurately model the modified takeoff conditions, changes were made to the segmental horizontal and vertical velocities, as well as the pathway of the hip.

4.2.5 Modifications to Segmental Velocities

The following steps were completed to modify the horizontal and vertical velocities of the thigh and shank segments.

1. The resultant velocity vector was calculated from the horizontal and vertical components.

$$v_{R_z} = \sqrt{v_V^{\,2} + v_H^{\,2}}$$

where

v_{Rz} is the resultant velocity of segment z
v_V is the vertical velocity component of segment z
v_H is the horizontal velocity component of segment z

2. The angle of the resultant velocity vector for segment z was calculated.

$$\theta_z = 180 - (\text{acos}\,(v_h/v_{Rz}))$$

where

θ_z is the angle of the resultant velocity vector for segment z relative to the left horizontal (the direction of movement)

3. With the magnitude of the resultant vector v_{Rz} kept constant, angle θ_z was increased or decreased according to the simulation test condition, and the new corresponding horizontal and vertical velocity components were calculated.

Horizontal component

$$V_{Hnew} = v_{Rz} \cos(\theta_z \pm x)$$

Vertical component

$$V_{Vnew} = v_{Rz} \sin(\theta_z \pm x)$$

where

V_{Hnew} is the new horizontal velocity
V_{Vnew} is the new vertical velocity
x is the change in angle based on the simulation test condition

4.2.6 Modifications to the Path of the Hip

When changes were made to the hip and knee angles at takeoff, it was anticipated that there would be corresponding changes seen in the parabolic pathway through which the athlete moves during flight. Evidence to support this was presented by Hunter et al. (2004) in their analysis of the relationship between step rate and step length. Vertical velocity was found to be a possible source of the negative relationship between step rate and step length, in which higher vertical velocities resulted in longer flight times and subsequently produced longer step lengths but slower step rates.

Mann (1985) reported that "good" male sprinters attain a mean vertical velocity of 0.52 m/s at takeoff, while "average" male sprinters reach 0.61 m/s, and "poor" male sprinters 0.69 m/s. Given that the participants in the present study were elite athletes, it is unlikely that they would demonstrate larger than ideal vertical velocities at takeoff. Based on Mann's data it would be plausible that elite sprinters show vertical velocity values between 0.52 and 0.60 m/s, a range of 0.08 m/s. For the purposes of this simulation study, this range was used to modify the vertical displacement of the hip, by increasing or decreasing the vertical velocity of the hip at takeoff by 0.04 m/s in association with the more extended (Hip+3 and Hip+6) or less extended (Hip-3 and Hip-6) hip angle. The more extended the hip, the lower the adjusted vertical velocity will be because the athlete is projecting themselves forward rather than upwards. Conversely, the less extended the hip, the higher the adjusted vertical velocity.

As an example, a vertical velocity at takeoff of 0.5 m/s would result in a vertical displacement during the flight phase of:

$v_f = v_i + 2ad$
$-2ad = v_i + 0$
$d = v_i / -2a$
$d = 0.5\ m/s / -(2 \times -9.81\ m/s^2)$
$d = 0.025\ m$

Repeating this method with varying vertical velocities gives:

0.54 m/s = 0.028 m
0.58 m/s = 0.030 m
0.46 m/s = 0.023 m
0.42 m/s = 0.021 m

These values are only for the flight phase. Because the vertical displacement is the total vertical displacement during both the flight and support phases (Hunter et al., 2004), these values will be doubled to produce the new adjusted vertical displacement values for the simulation conditions. This would produce vertical displacement values similar to the mean value of 0.05 metres presented by Mero, Luhtanen and Komi (1986) for elite sprinters.

To modify the horizontal displacement during the stride cycle, the formula presented by Hunter et al. (2004) for calculating the horizontal displacement during the flight phase was used. The flight distance may be determined by

$$D_{flight} = [v^2 \cdot \sin\theta \cdot \cos\theta + v \cdot \cos\theta \cdot ((v \cdot \sin\theta)^2 + 2 \cdot g \cdot h)^{0.5}]/g$$

where

v is the resultant velocity of the hip at takeoff
h is the difference in hip height between takeoff and touchdown
g is the gravitational constant 9.81 m/s^2

The modified vertical velocity and the horizontal velocity of the hip were used to determine the resultant velocity. The difference in hip height was obtained from the video data and remained constant for all trials.

4.3 Results

Tables 4-3 and 4-4 show the mean peak hip flexion and extension angular velocities through different takeoff angles at the hip and knee across all participants. For example, the mean peak hip flexion angular velocity when the hip angle was Hip+3 and the knee was Knee+3 was -862°/sec. Mean and standard deviations were calculated for each data column (across hip conditions when knee angle was kept constant) and row (across knee conditions when hip angle was kept constant). The value to which all others were compared was that corresponding to Hip 0 and Knee 0 which were the takeoff angles as measured from the sprinters as found in Project 1 and used in the initial simulation. The data shows that faster hip flexion angular velocities were seen when larger hip flexion angles (Hip+6) were coupled with more knee extension (Knee+6). The slowest hip flexion angular velocities were seen when the hip was less extended than usual (Hip-3 and Hip-6) and the knee was less extended (Knee-3 and Knee-6). Across hip conditions when the knee angle was kept constant (i.e., reading down columns), angular velocities were slower when the hip was less extended

(Hip-3 and Hip-6) and faster when the hip was more extended (Hip+3 and Hip+6) as compared to the Hip 0 condition. Across knee conditions when the hip remained constant (i.e., reading across rows), larger values were seen when the knee was more extended at takeoff (Knee+3 and Knee+6) and smaller when the knee was more flexed (Knee-3 and Knee-6).

Table 4-3. Comparison of mean peak hip flexion angular velocities from simulation.

	Knee-6	Knee-3	Knee 0	Knee+3	Knee+6	Mean	S.D.
Hip-6	-705	-729	-752	-779	-772	-747	31
Hip-3	-727	-749	-775	-796	-814	-772	35
Hip 0	-755	-781	-805	-821	-845	-801	35
Hip+3	-788	-810	-843	-862	-875	-836	36
Hip+6	-826	-855	-880	-904	-923	-877	39
Mean	-760	-785	-811	-832	-846		
S.D.	48	50	52	51	58		

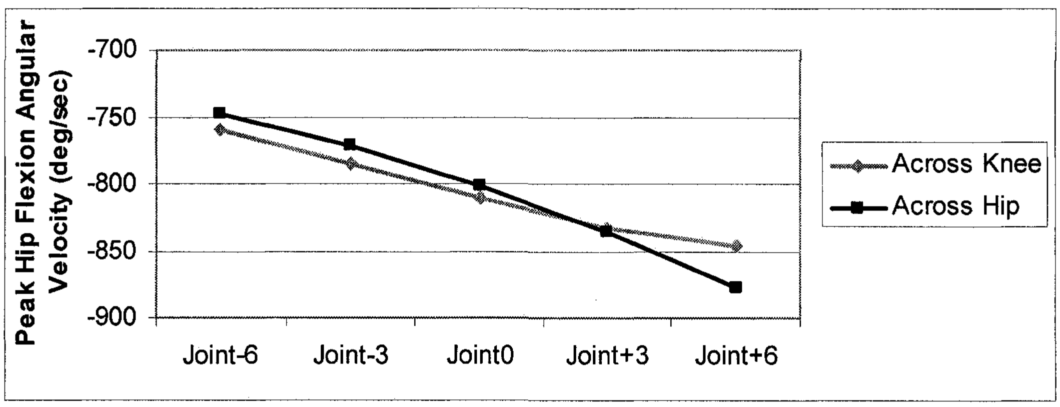


Figure 4-8. Mean peak hip flexion angular velocities across hip and knee conditions.

The fastest peak hip extension angular velocities were seen at the Hip+6 condition, with the highest value at Hip+6/Knee+6. The slowest extension angular velocities were seen when the hip was limited (Hip-6) and the knee was less extended (Knee-3 and Knee-6). Across hip conditions when the knee angle was constant, hip extension angular velocities were faster than sprinting when the hip was more extended (Hip+3 and Hip+6) at takeoff and comparable to sprinting when the hip was less extended (Hip-3 and Hip-6) (Figure 4-9). Standard deviations show that the variation among individuals in hip extension angular velocity was less than for hip flexion angular velocity.

Table 4-4. Comparison of mean peak hip extension angular velocities from simulation.

	Knee-6	Knee-3	Knee 0	Knee+3	Knee+6	Mean	S.D.
Hip-6	334	330	329	358	336	338	12
Hip-3	331	319	328	356	365	340	20
Hip 0	330	330	328	328	358	335	13
Hip+3	341	328	338	349	377	347	19
Hip+6	358	357	360	378	387	368	14
Mean	339	333	337	354	365		
S.D.	11	14	14	18	19		

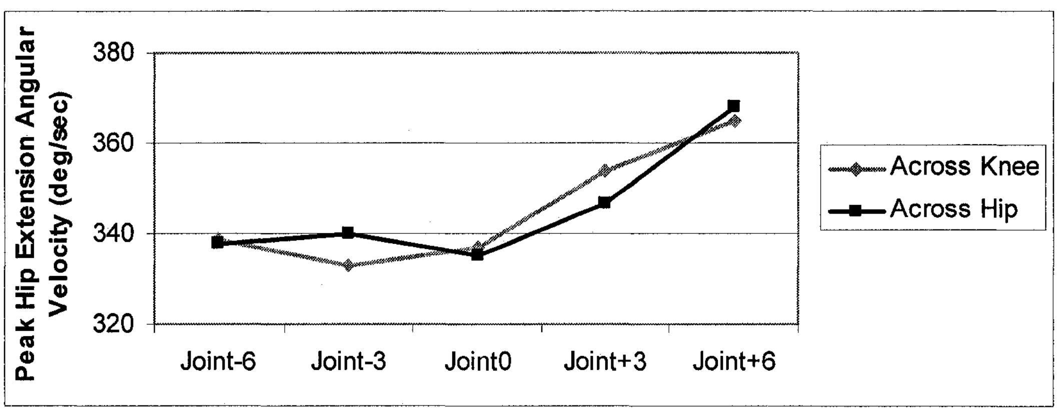


Figure 4-9. Mean peak hip extension angular velocities across hip and knee conditions.

Mean peak hip flexion angles corresponding to various hip and knee takeoff angles are seen in Table 4-5. The data is comparable to that of peak hip flexion angular velocity in that the largest hip flexion angles are seen when hip extension is increased at takeoff (Hip+6) and the smallest hip flexion angles when hip extension is the smallest at takeoff (Hip-6). Reading Table 4-5 across hip conditions, the hip reached a greater flexion angle when it was more extended at takeoff (Hip+3 and Hip+6) and a smaller flexion angle when the hip was less extended (Hip-3 and Hip-6). Across knee conditions, changes in hip flexion were smaller than when the hip angle was changed. Slightly smaller hip flexion angles were found when the knee was more extended at takeoff (Knee+3 and Knee+6) and larger when the knee was less extended (Knee-3 and Knee-6).

Table 4-5. Comparison of mean peak hip flexion angles from simulation.

	Knee-6	Knee-3	Knee 0	Knee+3	Knee+6	Mean	S.D.
Hip-6	139	144	148	153	153	147	6
Hip-3	140	144	149	153	156	149	6
Hip 0	142	147	151	154	158	150	6
Hip+3	145	149	155	158	159	153	6
Hip+6	148	152	157	161	164	156	7
Mean	143	147	152	156	158		
S.D.	3	4	4	4	4		

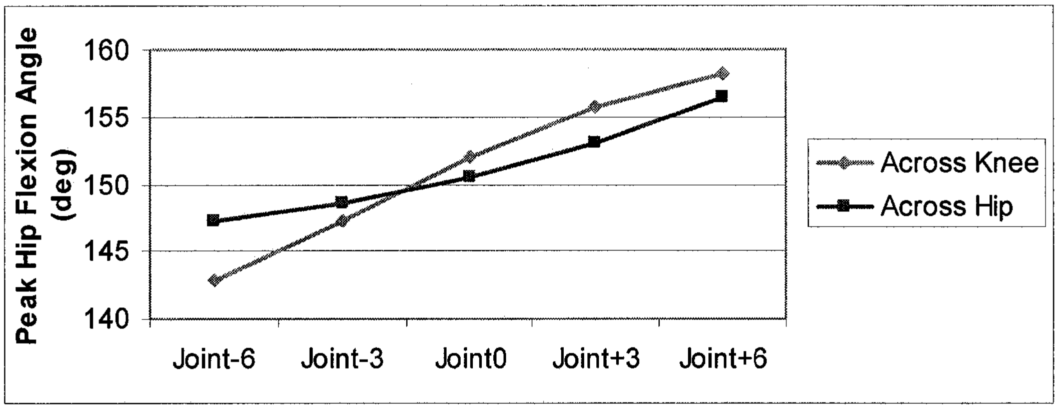


Figure 4-10. Mean peak hip flexion angles across hip and knee conditions.

To outline how the hip angular velocity curves changed with varying takeoff positions, graphs showing the curves across hip conditions with a constant knee angle (Knee 0) for subjects 2 and 14 are presented in Figures

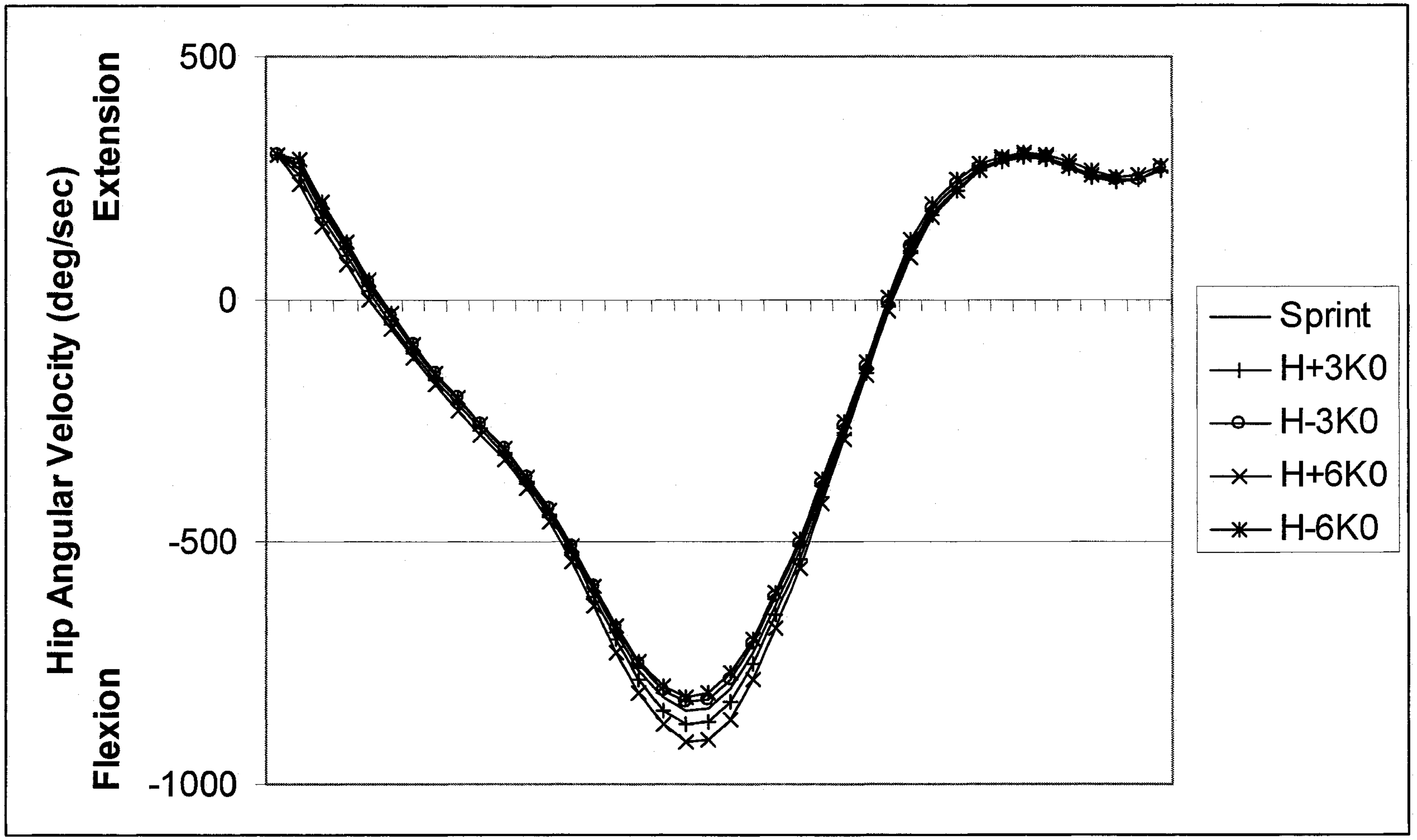


Figure 4-11. Hip angular velocity curves across hip conditions with a constant knee angle of Knee 0 for subject 2.

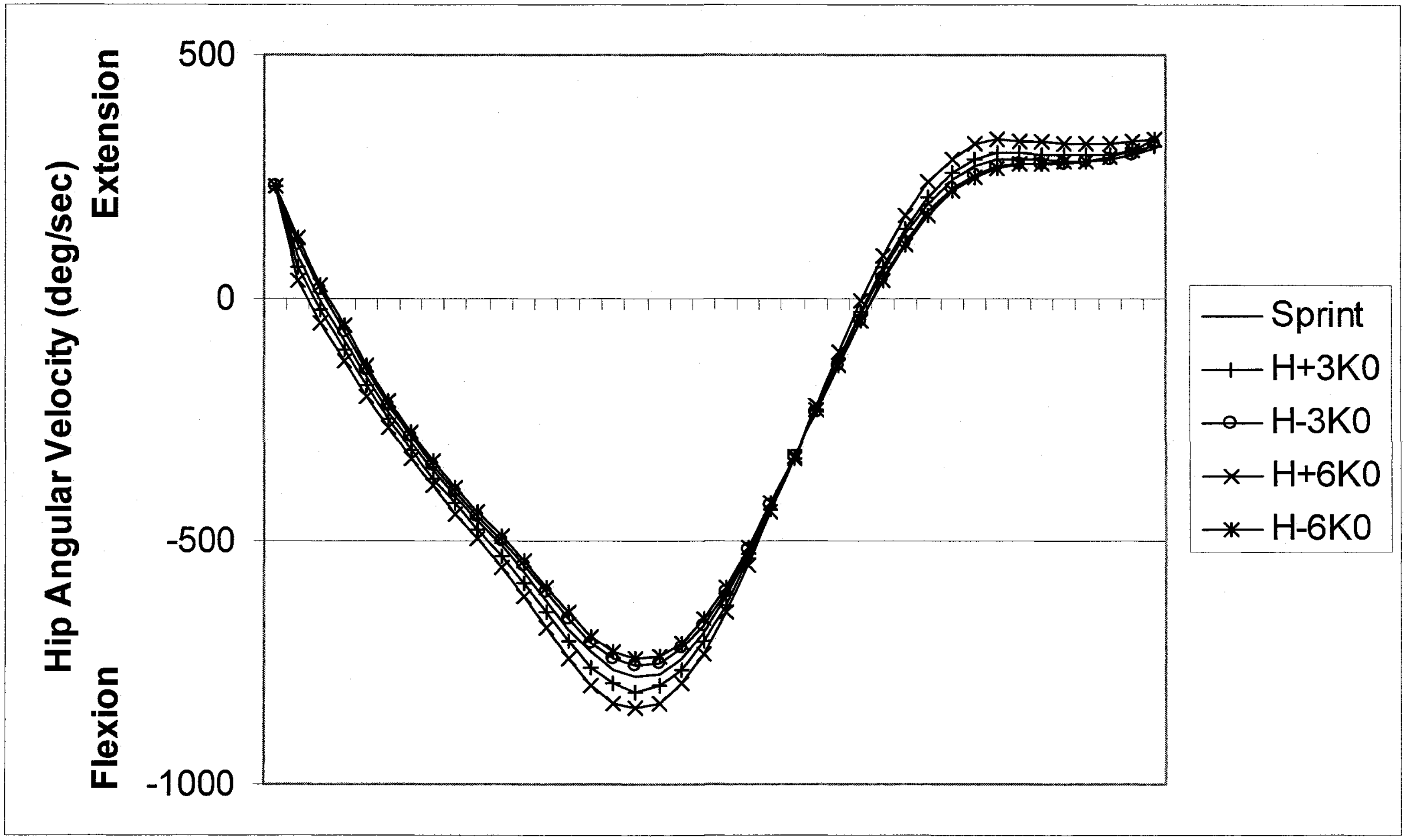


Figure 4-12. Hip angular velocity curves across hip conditions with a constant knee angle of Knee 0 for subject 14.

4-11 and 4-12. Both faster hip flexion angular velocities and hip extension angular velocities were seen at the Hip+3 and Hip+6 conditions as compared to sprinting.

At the knee, peak flexion angular velocities were fastest when the knee was more extended (Knee+3 and Knee+6) and the hip was less extended (Hip-3 and Hip-6) at takeoff (Table 4-6). The slowest knee flexion angular velocities were reported under conditions of limited knee extension (Knee-3 and Knee-6) and excessive hip extension (Hip+3 and Hip+6). When the knee angle was kept constant, Figure 4-13 shows that faster knee flexion angular velocities were seen when the hip was more extended (Hip+3 and Hip+6) and slower when the hip

Table 4-6. Comparison of mean peak knee flexion angular velocities from simulation.

	Knee-6	**Knee-3**	**Knee 0**	**Knee+3**	**Knee+6**	Mean	S.D.
Hip-6	945	986	1027	1073	1087	1023	60
Hip-3	965	1008	1042	1082	1123	1044	62
Hip 0	989	1026	1063	1092	1137	1061	57
Hip+3	1016	1046	1092	1120	1144	1084	52
Hip+6	1045	1080	1113	1145	1177	1112	52
Mean	992	1029	1067	1102	1134		
S.D.	40	36	35	30	33		

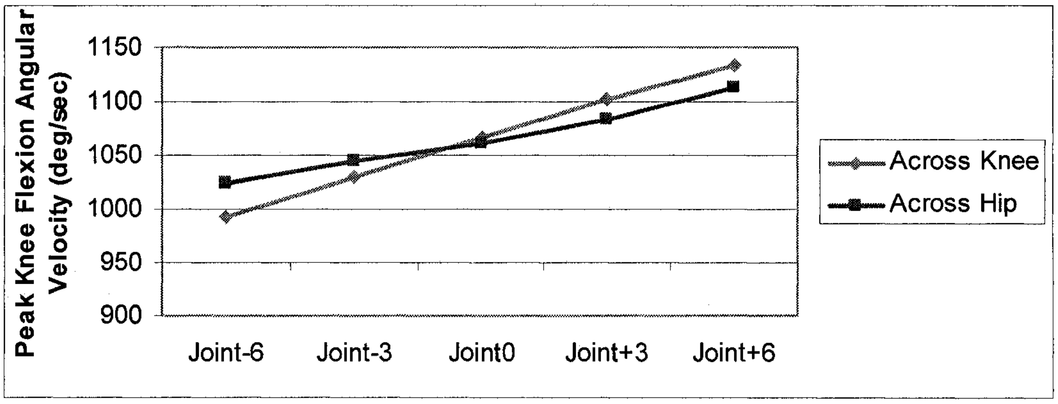


Figure 4-13. Mean peak knee flexion angular velocities across hip and knee conditions.

was less extended (Hip-3 and Hip-6). When hip angle was constant, more extended knee angles at takeoff (Knee+3 and Knee+6) resulted in knee flexion angular velocity that was faster than for sprinting. Conversely, when the knee was less extended at takeoff (Knee-3 and Knee-6), knee flexion angular velocities were slower than for sprinting.

In Table 4-7, mean peak knee extension angular velocities across hip and knee angles are reported. The fastest mean angular velocity was seen with Hip+6/Knee+6 at –1386°/sec. The slowest angular velocities were seen at the Hip-6/Knee-6. Across hip conditions when knee angle was constant, faster knee extension angular velocities were seen when the hip was more extended (Hip+3

Table 4-7. Comparison of mean peak knee extension angular velocities from simulation.

	Knee-6	Knee-3	Knee 0	Knee+3	Knee+6	Mean	S.D.
Hip-6	-1036	-1079	-1123	-1174	-1124	-1107	52
Hip-3	-1038	-1066	-1134	-1182	-1231	-1130	80
Hip 0	-1051	-1102	-1153	-1189	-1269	-1153	84
Hip+3	-1078	-1099	-1173	-1252	-1339	-1188	108
Hip+6	-1131	-1174	-1234	-1306	-1386	-1246	102
Mean	-1067	-1104	-1163	-1221	-1270		
S.D.	40	42	44	57	101		

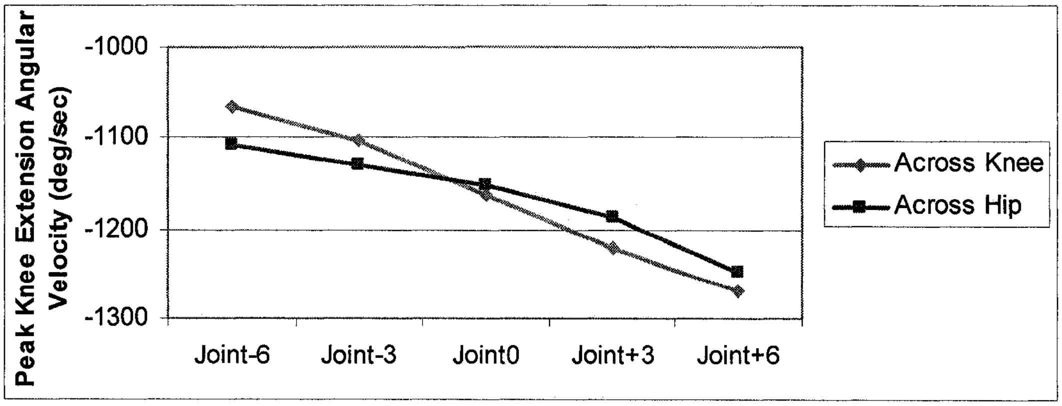


Figure 4-14. Mean peak knee extension angular velocities across hip and knee conditions.

and Hip+6) at takeoff, and slower knee extension angular velocities were found when the hip was less extended (Hip-3 and Hip-6) (Figure 4-14). Across knee conditions when hip angle was constant, the more extended the knee at takeoff (Knee +3 and Knee+6) the faster the knee extension angular velocities.

Examples of knee angular velocities across the five knee conditions with a constant hip angle (Hip 0) are presented in Figures 4-16 and 4-17. The fastest knee angular velocities, both in the flexion and extension directions can be seen when the knee is more extended at takeoff (Knee+3 and Knee+6).

In Table 4-8, the smallest mean minimum knee flexion angles were seen when greater hip extension (Hip+6) was coupled with increased knee extension

Table 4-8. Comparison of mean minimum knee flexion angles from simulation.

	Knee 0	Knee+3	Knee+6	Knee-3	Knee-6	Mean	S.D.
Hip-6	43	42	42	43	44	43	1
Hip-3	41	41	41	42	42	42	1
Hip 0	40	40	39	40	40	40	0
Hip+3	37	38	38	38	39	38	1
Hip+6	36	36	36	37	37	36	1
Mean	39	39	39	40	41		
S.D.	3	2	2	3	2		

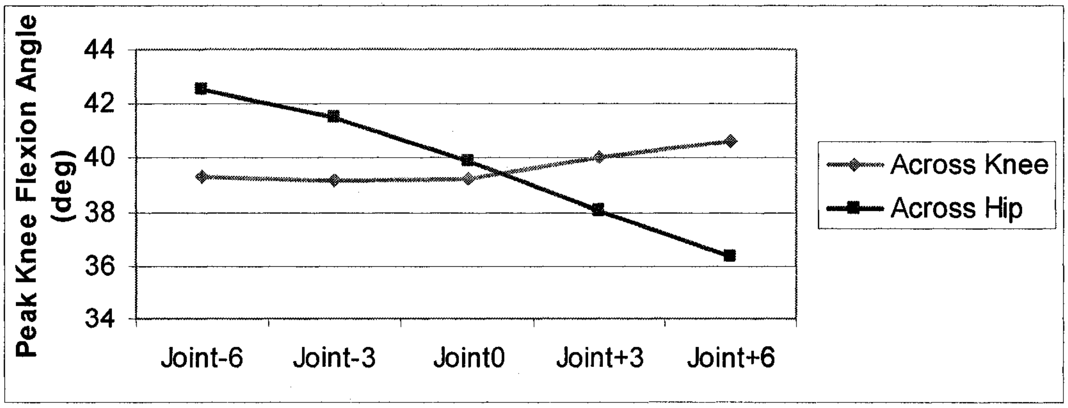


Figure 4-15. Mean minimum knee flexion angles across hip and knee conditions.

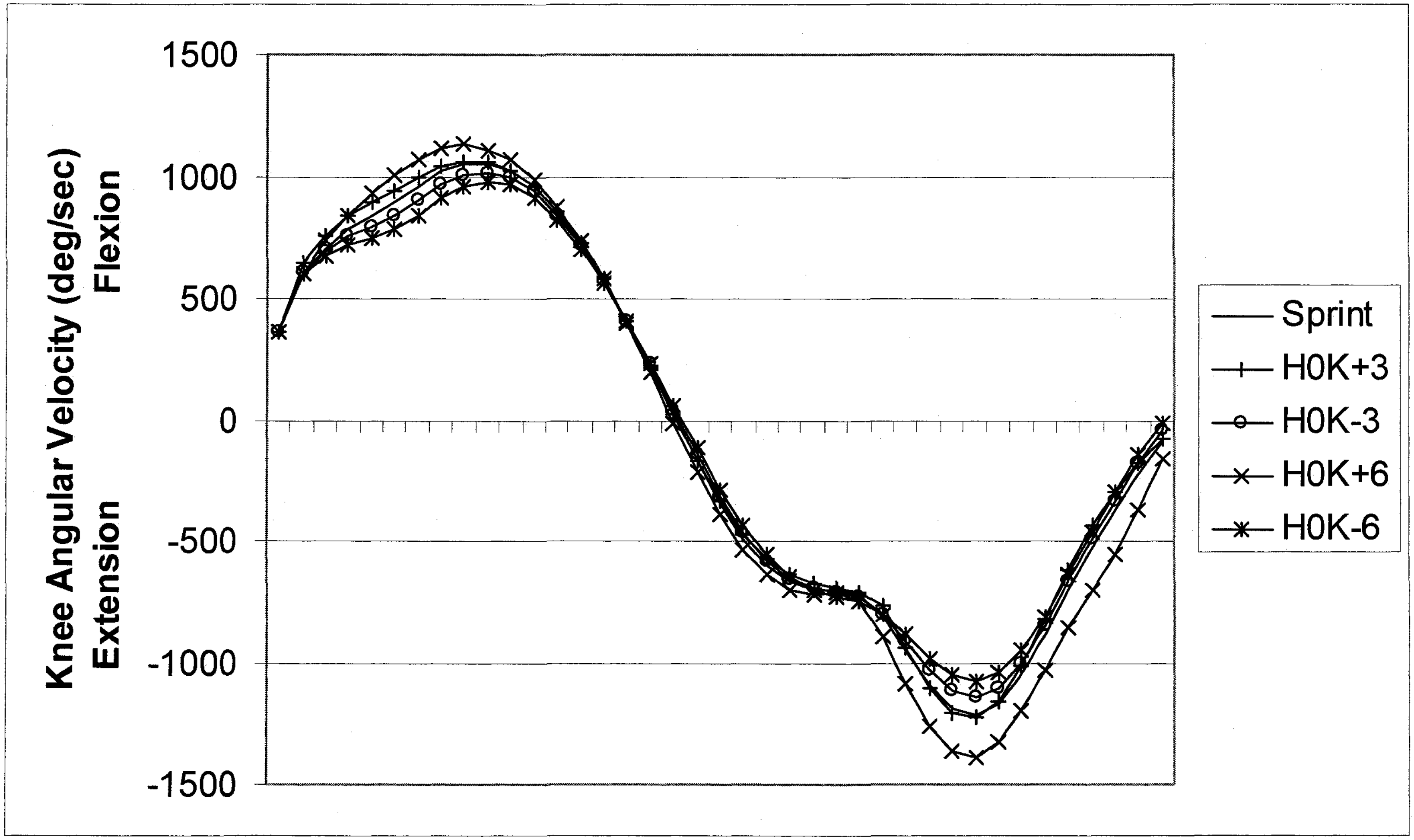


Figure 4-16. Knee angular velocity curves across knee conditions with a constant knee angle of Hip 0 for subject 2.

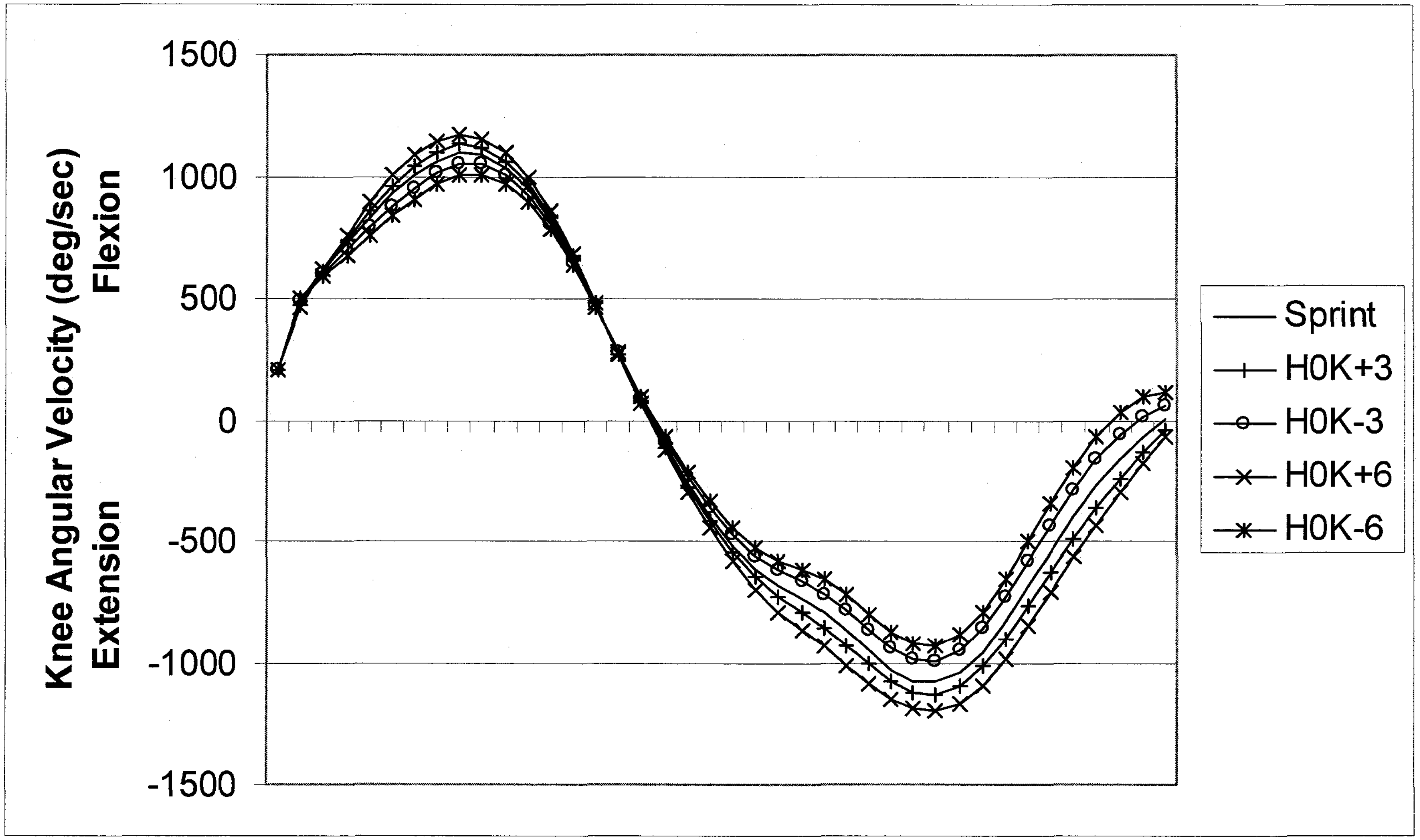


Figure 4-17. Knee angular velocity curves across knee conditions with a constant knee angle of Hip 0 for subject 14.

(Knee+3 and Knee+6). The largest mean minimum knee flexion angles were seen when the hip was less extended (Hip-6) and the knee extension was limited (Knee-3 and Knee-6). Across hip conditions when the knee angle was constant, limiting hip extension at takeoff (Hip-3 and Hip-6) reduced the amount of knee flexion, whereas greater hip extension at takeoff (Hip+3 and Hip+6) increased the amount of knee flexion (Figure 4-15). Across knee conditions when the hip angle was constant, minimum knee flexion did not vary appreciably when the knee angle changed at takeoff, as was seen by the small standard deviations.

Table 4-9 compares mean knee extension angles prior to touchdown across a range of hip and knee takeoff angles. Note that the knee extension angles at touchdown when the hip extension was increased (Hip+6) and knee extension was increased (Knee+6) at takeoff are greater than 180°, meaning the knee was hyperextended. Also, the knee extension angles at touchdown when the hip was limited (Hip-6) resulted in knee angles which may be flexed more than desired. Acceptable knee extension angles at touchdown were seen across the knee angles when the hip angles were not extreme.

The foot segment was included in the model in order to more accurately simulate the recovery leg during sprinting, as it was omitted in previous research (Philips et al., 1983; Wood et al., 1987). The computer model was run with similar adjustments to the ankle position at takeoff as were made for the hip and knee; however the results did not show any significant changes to the resulting hip and knee kinematics. The mass of the foot is relatively small in comparison to the other segments of the leg. Despite the fact that the foot segment is furthest away from the hip joint and its position could be a factor in the rotational inertia of the recovery leg, small angular changes of ankle position at takeoff do not affect the subsequent kinematics at the hip and knee.

Table 4-9. Comparison of mean knee extension angles prior to touchdown from simulation.

	Knee-6	Knee-3	Knee 0	Knee+3	Knee+6	Mean	S.D.
Hip-6	146	148	151	156	153	151	4
Hip-3	148	148	152	158	164	154	7
Hip 0	151	153	157	159	169	158	7
Hip+3	157	155	157	169	174	162	9
Hip+6	166	168	172	179	186	174	8
Mean	154	154	158	164	169		
S.D.	8	8	8	10	12		

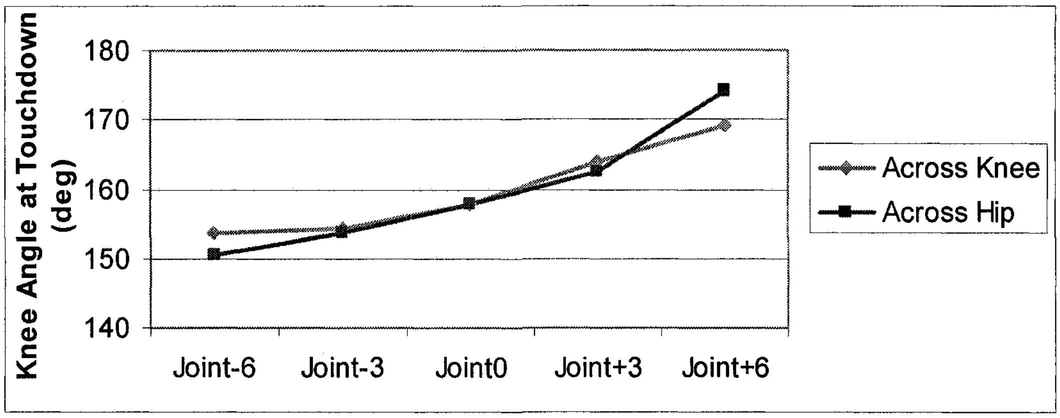


Figure 4-18. Mean knee extension angles prior to touchdown across hip and knee conditions.

Table 4-10 shows mean stride lengths as presented by the simulation model. Stride length was measured from takeoff to the point of contact of the same foot. The conditions which most closely replicated the sprint condition (Hip 0/Knee 0) were those when either the hip or the knee remained unchanged. The more the hip was extended at takeoff (Hip+3 and Hip+6) the longer the resulting stride length, whereas less hip extension at takeoff (Hip-3 and Hip-6) produced shorter stride lengths. In addition, greater knee extension (Knee+3 and Knee+6) was shown to increase stride length and less knee extension (Knee-3 and Knee-6) resulted in comparatively shorter stride lengths (Figure 4-19).

Table 4-10. Comparison of mean stride lengths from simulation.

	Knee-6	Knee-3	Knee 0	Knee+3	Knee+6	Mean	S.D.
Hip-6	4.39	4.49	4.59	4.71	4.66	4.57	0.13
Hip-3	4.46	4.53	4.67	4.77	4.88	4.66	0.17
Hip 0	4.55	4.65	4.76	4.84	4.99	4.76	0.17
Hip+3	4.66	4.71	4.85	4.94	5.06	4.84	0.16
Hip+6	4.82	4.90	4.99	5.09	5.19	5.00	0.15
Mean	4.57	4.65	4.77	4.87	4.95		
S.D.	0.17	0.16	0.16	0.15	0.20		

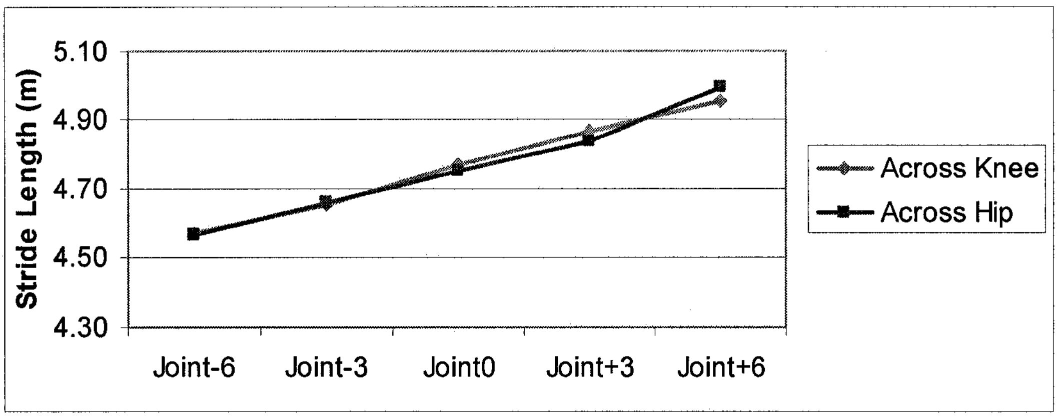


Figure 4-19. Mean stride lengths across hip and knee conditions.

4.4 Discussion

The software used in this investigation, Working Model 2D, has been used previously in modeling and simulation studies involving gait and locomotion. Yakovenko, Gritsenko, and Prochazka (2004) used Working Model 2D in their simulation study examining the contribution of stretch reflexes to locomotor control. They used a two-legged planar model with nine segments and which were driven by 12 musculotendonous actuators. The model's structure was based on the hindlimb of the cat. Ferris, Liang, and Farley (1999) used this software in their analysis of how runners change their leg stiffness when encountering a new surface. They developed a spring-mass model with input parameters collected from both force plate data and from high speed video. The advantage of using this software is that it allows the investigator to visualize the model being developed and the simulation being run. The model can be constructed from a variety of geometric shapes, joints, and controls and constraints can be added to or removed from the system when desired. To run the simulation, data may be read via DDE from a program such as Microsoft Excel or Matlab. Output data from the simulation may also be exported to a spreadsheet file for further analysis.

Simulation models vary in their level of sophistication and complexity, ranging from simple models (i.e. Farley & Gonzalez, 1996) to very complex models (i.e., Hatze, 1981). The model used in this study was a torque driven model with the inputs obtained from experimental data. Models with similar levels of sophistication have been used previously in simulation and optimization studies involving human gait and running (Phillips, Roberts, & Huang, 1983; Wood, Marshall, & Jennings, 1987; Onyshko & Winter, 1980; Mena, Mansour, & Simon, 1981). Although the current model does not reach the level of sophistication and complexity of some models, it was appropriate to answer the research question posed. Hubbard (1993) stated that one should always begin with a model which has the simplest design yet still captures the essence of the question being investigated. From this simplified model, additional complexity should be added only when it is necessary. Alexander (1992) believed that is not always necessary for models to be developed with as much complexity as possible. Simple models may at times not be appropriate, but it is generally advisable to keep the model as simple as is consistent with its task.

Optimal control theory has been applied when modeling and simulating human gait, often with a criterion of minimum energy (Beckett & Chang, 1968; Chow & Jacobson, 1972; Nubar & Contini, 1961; Seirig & Arvikar, 1973). Mena, Mansour, and Simon (1981) believed that optimal control studies may be mathematically elegant, but their focus has been on the optimal nature of human gait rather than examining the observable and alterable physical variables which produce the overall movement pattern. In the optimization study of Wood et al. (1987) one of the objective functions was to minimize recovery time. The results showed that less hip flexion and greater hip extensor activity would produce the optimal movement pattern. Theoretically this may be so, however, these changes may not be feasible for the sprinter to make. Shortening the stride of one leg will result in changes to the other leg as well in order for the two legs to continue moving synchronously. Although the results from Wood et al. (1987) provided some valuable insight into optimizing sprinting technique, the results must be discussed in relation to the entire recovery phase or the entire sprint stride. Simulation models such as the one developed for this study serve an important function in understanding how changing one aspect of the movement will affect others in practical terms and not based on a theoretical optimum alone.

When modeling with Working Model 2D, or with any other modeling software program, there are certain assumptions which are made. It was assumed that the model's segments were rigid bodies of uniform mass distribution, and that the joints were frictionless hinge joints. Also, it was assumed that the motion being simulated, that is, the sprinting stride was a planar action with all segments moving in the anterior-posterior direction. There were also assumptions specific to the model developed for this project. When modifications were made to the path of the hip in association with the changes in hip and knee angles at takeoff, the methods that were used to determine the modified hip path had been used previously for measuring the horizontal displacement of the centre of mass. It was assumed that the hip joint and centre of mass follow similar paths.

As described by Whittlesey and Hamill (2004), there are limitations to using computer models to simulation motion. The human system is far more complicated than any model which could be used to represent it, and it is not possible to control for all variables, therefore assumptions must be made. Also, one must limit the amount of generalization from the results because the model is driven by resultant joint moments, not by individual muscles. There must be a balance between theory and application.

After takeoff, developing fast hip flexion angular velocity is advantageous as it will allow the recovery leg to swing forward faster and enable the sprinter to achieve a better peak hip flexion position (Mann, 1985). Larger hip flexion angles will then allow the sprinter to develop a faster hip extension angular velocity and an opportunity to contact the ground in a more optimal position (Mann, 1985). Faster hip flexion angular velocities and greater hip flexion angles were seen with increased hip extension (Hip+3 and Hip+6). This is consistent with the significant correlation found in Project 1 between maximum hip extension angle and knee flexion angular velocity extension ($r = -0.703$, $p < 0.01$). The more extended the hip at takeoff, the faster the knee flexes which will allow the leg to achieve a more flexed position and swing through faster. The results also show that hip position at takeoff was more influential on peak hip flexion than knee position at takeoff. When the hip angle was kept constant and the knee angle was changed (i.e., reading across rows in Table 4-5) the resulting change in hip flexion was not as dramatic as was seen at constant knee angles with changing hip angles (i.e., reading down columns in Table 4-5). The importance of the action at the hip during gait has been shown by Mena, Mansour, and Simon (1981) in their simulation of the swing phase of gait in which they reported the overall dynamics of the swing leg were most dependent upon the initial angular velocity of the thigh. Mann and Sprague (1983) stated that the greatest contributor to sprint success is the muscular activity at the hip. From toe-off to footstrike, the hip flexor activity needed to recover the leg, as well as the hip extensor effort produced to halt leg recovery and extend the limb toward footstrike are significant contributors.

Figures 4-20 and 4-21 show photosequences of the computer simulation at the Hip-6/Knee 0 and Hip+6/Knee 0 conditions for subject 3. Early in recovery as the thigh began to swing forward and the knee flexed, the increased hip extension condition (Hip+6/Knee 0) resulted in a faster peak knee flexion angular velocity and smaller minimum knee angle (Figure 4-12 photo 2 and 3). The result was a decrease in the moment of inertia of the recovery leg which enabled it to swing through with a faster hip flexion angular velocity. With the leg swinging forward faster there was a corresponding increase in peak hip flexion angle. Phillips, Roberts, and Huang (1983) found similar results in their simulation of the swing leg in that as the hip flexion angular velocity increased there was a corresponding increase in knee flexion angular velocity.

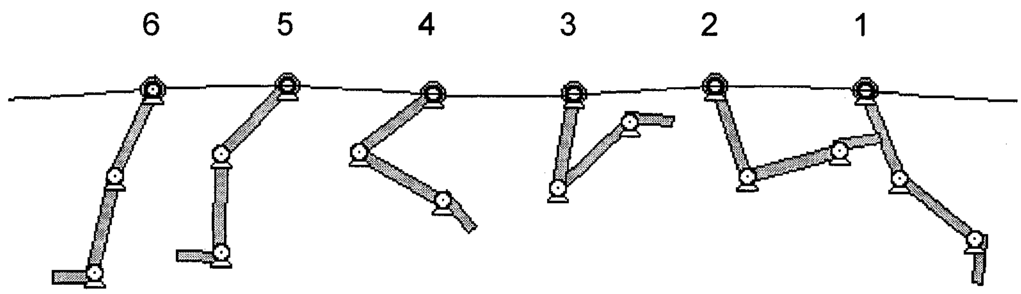


Figure 4-20. Photosequence of the Hip+6/knee 0 condition for subject 3.

In contrast, the less extended hip position in Figure 4-13 resulted in a delay in the flexion of the knee (Figure 4-13 photo 2), which prevented the recovery leg from swinging through fast enough to produce an adequate hip flexion angle to allow for proper ground contact. The result was a limited stride length with the foot contacting under the body.

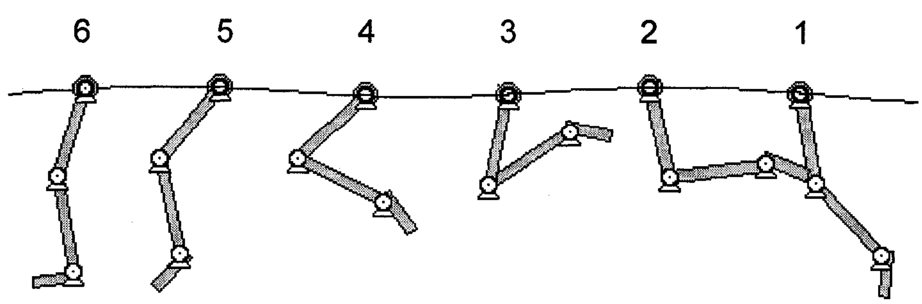


Figure 4-21. Photosequence of the Hip-6/Knee 0 condition for subject 3.

Late in the recovery phase as the hip was extending to bring the foot to the ground for support, faster hip extension angular velocities were seen when the hip was more extended at takeoff. From this takeoff position there was an increased knee flexion angular velocity which resulted in a smaller minimum knee angle and a faster hip flexion angular velocity. However, this minimum knee angle also meant that the knee was less extended when the hip reached the maximum flexion angle and began to extend. This resulted in a decreased moment of inertia during extension which produced a faster hip extension angular velocity.

Decreasing the moment of inertia during the forward swing of the recovery phase by flexing the hip and knee has been discussed previously (Hay, 1993; Hall, 1995), however, there has been no reference to using this same concept during the latter phase of recovery as the hip and knee are extending. This idea could be an important factor in improving sprinting performance as a reduction of the moment of inertia during hip extension would allow the hip to extend faster which would decrease the relative velocity of the foot prior to ground contact and reduce the braking force at contact.

At the onset of the recovery phase, Bunn (1978) and Dillman (1975) stated that one of the most common errors in sprinters was incomplete knee extension at takeoff and after takeoff. In elite sprinters, Mann (1985) found that reducing knee extension at takeoff helps to minimize ground contact time. After takeoff, the muscle moment at the knee is dominated by the extensors which are working eccentrically to reduce the backwards angular momentum of the lower leg and foot (Dillman, 1971). The positions of the hip and knee at takeoff are factors in the degree to which the knee flexes. The smallest maximum knee flexion angle was seen when both the hip and knee were more extended. This is closely related to the significant correlation found in Project 1 between maximum angle of hip extension and peak knee flexion angular velocity ($r = -0.703$, $p < 0.01$). In addition, because the knee flexes passively during sprinting (Simonsen, Thomsen, & Klausen, 1985), the faster the peak hip flexion angular velocity the smaller the minimum knee flexion angle. A significant correlation was also found between these two variables in Project 1 ($r = 0.578$, $p < 0.05$).

During recovery, the hip angle at takeoff was more influential in determining the amount of knee flexion than knee angle at takeoff. Peak knee flexion angle only changed a small amount across knee conditions when hip angle at takeoff remained constant, whereas larger changes were seen across hip conditions when knee angle at takeoff was kept constant. Lemaire and Robertson (1990) stated that the role of the knee musculature during sprinting is primarily one of shock absorber and that there is virtually no contribution from the knee to the speed or height of the leg during recovery. The hip, in comparison, is responsible for most of the motion of the leg during sprinting, driving the leg through and back. The exception seen in this study was at Knee+6 in which larger knee flexion angles were found across all hip angles at takeoff. The more extended the knee is at takeoff the less chance there is for the knee to flex adequately, regardless of what the hip angle is.

Faster knee flexion angular velocities will allow the recovery leg to achieve a flexed position earlier and will reduce the moment of inertia and resistance to forward rotation. Sprint literature also suggests that elite sprinters limit the amount of knee extension at takeoff in order to reduce ground contact time (Mann, 1985). The results of the computer simulation showed that fastest knee flexion angular velocities were seen when the knee was more extended at takeoff (Knee+3 and Knee+6), coupled with an increased hip angle (Hip-6). This same hip/knee combination, however, also resulted in the fastest knee extension angular velocity. This hip/knee action produced a sequential movement to "whip" the lower leg forwards (Wood et al., 1987). Knee extension late in the swing phase is facilitated without knee extensor musculature involvement when the speed of the thigh forward rotation decreases (Phillips et al., 1983). These results are supported by those of Project 1 in the correlations related to the knee through recovery. Sprinters who are more extended at the knee at takeoff flex the knee faster ($r = 0.55$, $p < 0.05$) which results in increased flexion angles, but this also leads to the knee extending at a greater rate ($r = 0.581$, $p < 0.05$). The resulting motion would produce knee angles which are beyond the physiological limitations of the knee (Figure 4-22). This result provides further support to the belief that the eccentric muscle moment at the knee prior to footstrike is a possible limiting factor in maximum sprinting speed (Chapman & Caldwell, 1983; Wood, 1987). Larger muscle moments than can be produced are required to control knee extension and initiate knee flexion.

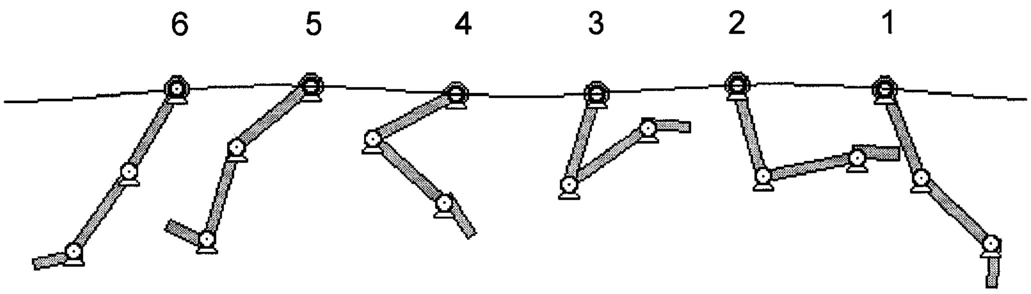


Figure 4-22. Photosequence of the Hip+6/Knee+6 condition for subject 4 resulting in physiologically impossible knee angles late in the recovery phase.

Sprinting speed is a function of stride length and stride rate. Initial increases in speed by an experienced runner are a result of an increased stride length. After a step of optimal length has been attained, further increases in speed become a matter of increasing stride frequency (Adrian & Cooper, 1989). Luhtanen and Komi (1978) found that at higher velocities stride length levelled off, whereas stride rate continued to increase. The results of this investigation show that it is possible for a sprinter to alter stride length by adjusting the angles at the hip and knee at takeoff. Changing the hip and knee angle at takeoff did not result in any differences in the takeoff distance as described by Hay (1993), meaning the increase or decrease in stride length was seen in the landing distance. Also, the simulation demonstrates that changes in hip angle are more influential on stride length than knee angle, as larger changes in stride length were seen across hip conditions as compared to across knee conditions. However, the results also suggest that simply changing the hip and knee takeoff angles will likely not result in faster sprint times by increasing stride length, as changes in takeoff position will also influence the landing position. If ground contact is too far in front of the centre of mass the braking force on contact will be too large and the sprinter will slow down with each stride. Conversely, if the foot contacts the ground underneath the sprinter contact time will be reduced and the athlete will not be on the ground long enough to properly change the direction of the centre of mass for the next flight phase and the range of motion of the entire stride will be shortened.

4.5 Conclusions and Implications for Training

The results of this simulation showed that the kinematics of the recovery leg during sprinting could be altered by changing the hip and knee positions at takeoff. When the hip was more extended and the knee was straighter at takeoff, there was a faster knee flexion angular velocity which resulted in greater knee flexion angle. With the moment of inertia reduced, the recovery leg swung forward faster with the hip reaching a greater flexion angle. This greater knee flexion angle also meant the knee was still more flexed when the hip began to extend, which produced a faster hip extension angular velocity because of the reduced moment of inertia. This increased hip extension angular velocity resulted in an increased knee extension angular velocity. Conversely, a less extended hip position at takeoff meant a slower knee flexion angular velocity and less knee flexion. The recovery leg did not swing forwards with the same angular velocity and the hip did not reach the same angle of flexion. There was also a decrease in the knee extension angular velocity which meant the knee was less extended when the hip reached its maximum flexion angle and began to extend. This resulted in an increased moment of inertia during extension which produced a slower hip extension angular velocity.

The hip and knee angles which produced the fastest hip flexion angular velocities and greatest hip flexion were the same as those with the fastest hip extension angular velocities. These takeoff positions, however, also resulted in extremely high knee extension angular velocities and produced knee angles which are beyond anatomical limits. It may be possible to improve in one area, but it will be to the detriment of another. In order to maximize performance, sprinters must find a compromise among these variables. The relative importance of the hip versus the knee in altering the kinematics of the recovery phase was also discussed, in which changes at the hip at takeoff are more influential on the recovery leg than at the knee. This supports the belief that the focus of training should be on the development of the musculature which controls hip movement rather than the knee joint.

In general, the results of the computer simulation were as hypothesized. The one area in which the results were somewhat surprising was in the action at the knee late in recovery, in which anatomically impossible knee angles resulted from limited hip angles and extended knee positions. If anatomical constraints were put on the model this important finding would not have been evident, but it does provide support for the limiting nature of the eccentric knee extensor activity during sprinting (Chapman & Caldwell, 1983).

The results of the computer simulation showed that changes in recovery leg kinematics can result from small modifications in hip and/or knee angle at takeoff. With proper training, including lower extremity strength and power development and using sprint drills to modify technique, it is feasible for these changes to be implemented. Changes of this small magnitude may not be as important when dealing with non-elite sprinters who have not matured physically or technically, however, they could be very important to improving sprinting success at the elite level.

This analysis provides important insight into the effect changing a sprinter's hip and knee positions at takeoff would have on the subsequent recovery kinematics. The results show what the benefits would be of such changes, but they also reveal what the negative consequences would be as well. It is important for athletes and for coaches to know this type of information when making changes to an athlete's sprint stride. For example, if a coach wanted to increase the degree of knee flexion during recovery, this change can be accomplished by limiting hip extension and knee extension but will result in the knee extending at a greater rate than may be desired. If the results of this study were to be applied to an athlete, it would not be advisable to have the athlete attempt to run at maximum speed and change their takeoff position. If this information was implemented at a lower speed such as a slow run and developed over time with proper guidance and training, it is possible that beneficial changes could result to the athlete's sprinting mechanics.

CHAPTER 5

Summary and Recommendations

The purposes of this study were twofold: (a) to determine what are the relationships among the kinematic and kinetic variables of the recovery phase of elite sprinters; and (b) to use computer simulation to determine what effect changing hip and knee angles at takeoff would have on the subsequent kinematics of the recovery leg. Project 1 was a kinematic and kinetic analysis of the recovery leg of elite sprinters. High speed video was collected during competition of elite male sprinters at maximum velocity. The results showed that there are relationships among the kinematic and kinetic variables associated with the recovery leg. The amount of knee extension at takeoff was correlated with the rate of knee flexion. Also, greater knee flexion angles were correlated with faster hip flexion angular velocities, and slower horizontal foot velocities were seen when the hip was flexed to a greater degree. Knee extensor moments and knee extension angular velocities were found to be related to hip flexion angular velocity as well. In addition, a number of variables were found to be related to both stride rate and stride length.

Project 2 was a computer simulation of the recovery leg during sprinting using input data from the 14 subjects Project 1. The results showed that the kinematics of the recovery leg during sprinting could be altered by changing the hip and knee positions at takeoff. When the hip was more extended and the knee was straighter at takeoff, there was a faster knee flexion angular velocity which resulted in greater knee flexion angle. With the moment of inertia reduced, the recovery leg swung forwards faster with the hip reaching a greater flexion angle. This greater knee flexion angle also meant the knee was still more flexed when the hip began to extend, which produced a faster hip extension angular velocity because of the reduced moment of inertia. This increased hip extension angular velocity resulted in an increased knee extension angular velocity. Conversely, a less extended hip position at takeoff meant a slower knee flexion angular velocity and less knee flexion. The recovery leg did not swing forward with the same angular velocity and the hip did not reach the same angle of flexion. There was also a decrease in the knee extension angular velocity which meant the knee was less extended when the hip reached maximum flexion angle and began to extend. This resulted in an increased moment of inertia during extension which produced a slower hip extension angular velocity.

It is evident from these results that the variables associated with the recovery phase are closely related. Both studies showed that the leg position at takeoff dictates how it flexes, and that the degree of knee flexion influences how fast the leg swings forward. One of the most important findings that had not been previously discussed in the sprint literature is the relationship among the degree of knee flexion (moment of inertia), knee extensor moment, and knee extension angular velocity. The limiting factor in sprinting speed occurs when the sprinter is unable to develop sufficient eccentric muscle moment at the knee prior to ground contact (Chapman & Caldwell, 1983). Project 1 identified the significant relationship between knee extensor moment and the moment of inertia of the leg, and between the degree of knee flexion and knee extension angular velocity. Project 2 suggested that less knee flexion during recovery reduces the rate of knee extension, which could possibly reduce the knee flexor moment required to control knee extension. Further research is warranted to determine if this is practically feasible.

It is important to note that key results in both studies were found throughout the entire recovery phase. In particular, the significance of takeoff was made evident. Hip and knee angles at takeoff have been discussed previously from a performance point of view (e.g., Mann, 1985), but their importance in dictating subsequent leg movements has not previously been addressed. This early portion of recovery had also been omitted in previous computer simulation studies (Wood et al., 1987; Philips et al., 1983) which may limit the applicability of their findings. The motion of a body can vary greatly depending on its initial position (Chou, Song, & Draganich, 1995), and understanding how small changes can result in dramatically different movement patterns is crucial when trying to maximize performance.

Many studies on the biomechanics of sprinting have attempted to determine the “ideal” mechanics. Research has shown that there are kinematic and kinetic differences between athletes of different abilities (e.g., Mann, 1985; Vardaxis & Hoshizaki, 1989), but one must also recognize that there are also differences among athletes of similar abilities. The current research indicates that it is not necessary for sprinters to have “ideal” recovery leg mechanics. The participants involved in Project 1 were all elite athletes, and despite the fact that they demonstrated a range of techniques they still moved their foot from behind their body to in front in preparation for the next ground contact very effectively. It is important that coaches, athletes, and scientists have an understanding of the interaction among the variables associated with recovery, and that both strength and technique are considered when determining if changes should be made to an athlete's sprint mechanics.

This study included an analysis of 14 high level sprinters and incorporated a combination of statistical and simulation methods to determine relationships among variables related to performance and to determine the effects of modifying technique. These same methods could also be used to individualize a sprinter's training program as well. Once quantitative data is collected for an athlete and an assessment is made of their mechanics, simulation methods could be employed to prescribe modifications in order to improve performance. This would allow potential changes to first be examined using computer simulation in order to determine their effect on the subsequent leg movements and help reduce the risk of injury.

Based on the relevant literature and the research completed for this study, the following are recommended as future research objectives:

1. Further research is required in the biomechanics of sprinting. In particular, it is important to understand the variations in sprint technique among athletes of different ability levels more clearly, especially at the elite level. Differences between sprinters of different abilities have been investigated previously, but there has been no study into the intrasubject variability in the kinematics and kinetics of sprinting.

2. The concept of moment of inertia of the recovery phase should be investigated further. This study showed that moment of inertia plays an important role in not only the forward swing phase of the recovery leg, but also suggested its influence during the backward swing of the leg prior to ground contact.

3. Further research is needed in the modeling and simulation of sprinting. The effect of changing the recovery mechanics on the ground reaction forces produced during ground contact, which has been shown to be an important factor in successful sprinting, requires additional study.

4. The model developed for this study should be used for further investigation. Adjustments could be made to the joint moments and the resulting changes in the lower extremity kinematics could be examined. Also, this model could be used to study how changes in the lower extremity anthropometrics, as seen during child and adolescent development, could influence the leg movements during locomotion.

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Appendix 1

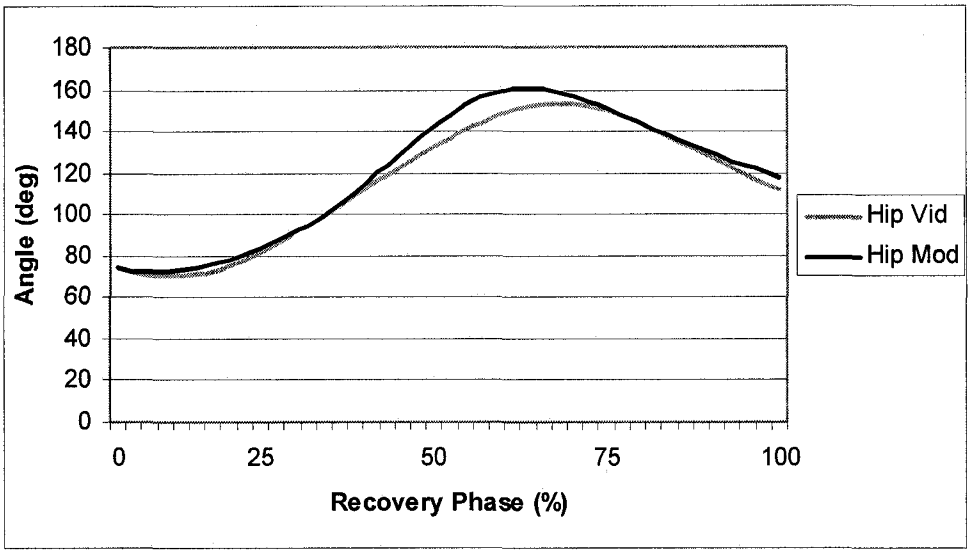
Individual Kinematic and Kinetic Data For Subjects in Project 1

Subject	Hip Angle at TO	Knee Angle at TO	Max Angle Hip Extension	Minimum Knee Angle	Peak Knee Flexion Ang Vel	Peak Hip Flexion Ang Vel	Max Angle Hip Flexion	Peak Knee Extension Ang Vel	Peak Hip Extension Ang Vel	Peak Hip Flexor Moment	Peak Hip Extensor Moment	Peak Knee Extensor Moment	Peak Knee Flexor Moment	Foot Velocity at Touchdown	Moment of Inertia of Leg	Stride Length	Stride Rate	100 m Time
1	74	150	70	47	1072	-590	156	-900	451	-258	246	-107	142	1.31	1.52	4.98	2.31	10.29
2	67	153	61	46	1087	-758	149	-976	535	-289	279	-113	139	3.29	1.28	4.72	2.45	10.29
3	74	140	71	37	827	-785	154	-1235	444	-287	255	-124	141	2.21	1.19	4.80	2.45	10.40
4	69	145	64	35	1221	-850	155	-1003	415	-325	254	-114	138	1.26	1.12	4.72	2.45	10.73
5	67	151	61	40	1355	-735	151	-923	549	-268	280	-133	167	1.96	1.35	4.63	2.45	10.83
6	78	151	63	37	1475	-650	152	-1112	445	-257	266	-110	139	1.85	1.27	4.73	2.35	10.64
7	75	144	71	56	1089	-589	154	-926	469	-254	232	-104	138	1.33	1.79	5.09	2.26	10.24
8	80	137	73	45	1033	-594	154	-941	422	-271	271	-114	157	1.86	1.39	4.84	2.35	10.48
9	69	149	60	32	1327	-740	153	-1381	442	-293	272	-129	136	2.49	1.11	4.87	2.35	10.58
10	79	150	65	45	1141	-827	164	-978	612	-301	246	-117	128	0.89	1.39	5.21	2.31	9.94
11	78	150	63	39	1072	-819	160	-1056	508	-239	242	-118	140	2.43	1.11	5.05	2.31	10.31
12	71	136	70	42	1057	-788	157	-1079	500	-259	309	-133	128	2.22	1.07	4.84	2.50	9.85
13	73	144	68	40	1079	-798	160	-1134	432	-268	275	-109	150	1.50	1.16	5.03	2.40	10.12
14	72	143	68	47	1047	-683	158	-1207	454	-299	302	-114	149	0.89	1.54	5.12	2.35	9.82

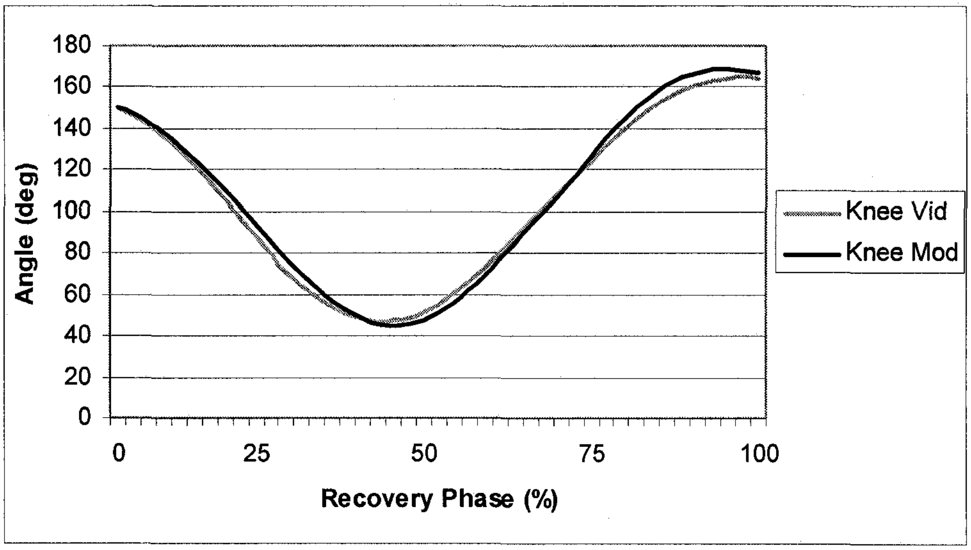
Appendix 2

Computer Simulation Output Data For Each Subject in Project 2

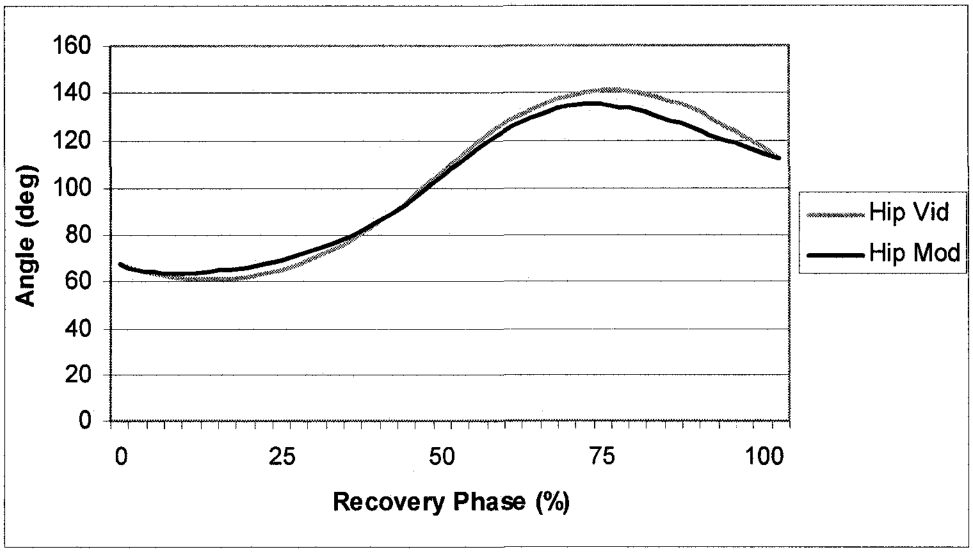
Subject 1 – Hip angular displacement data – video and model



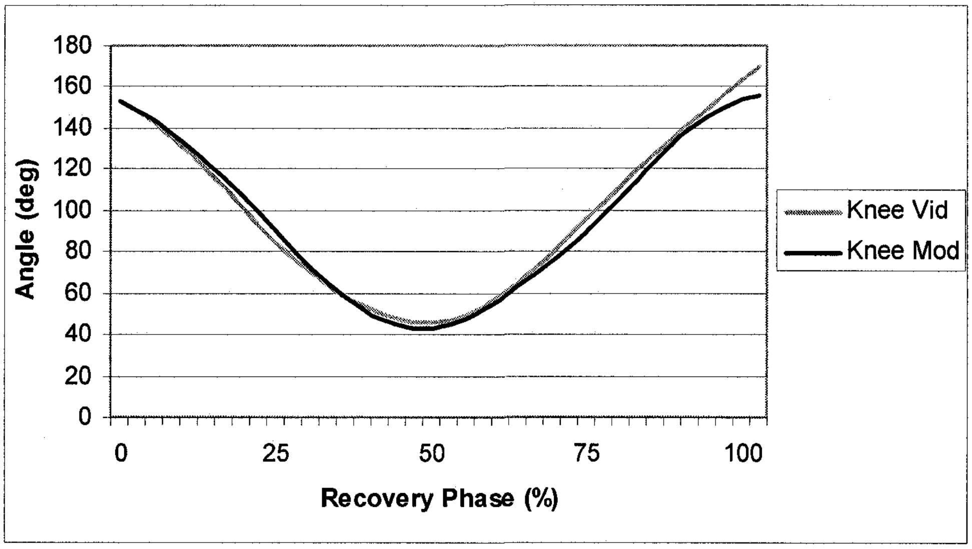
Subject 1 – Knee angular displacement data – video and model



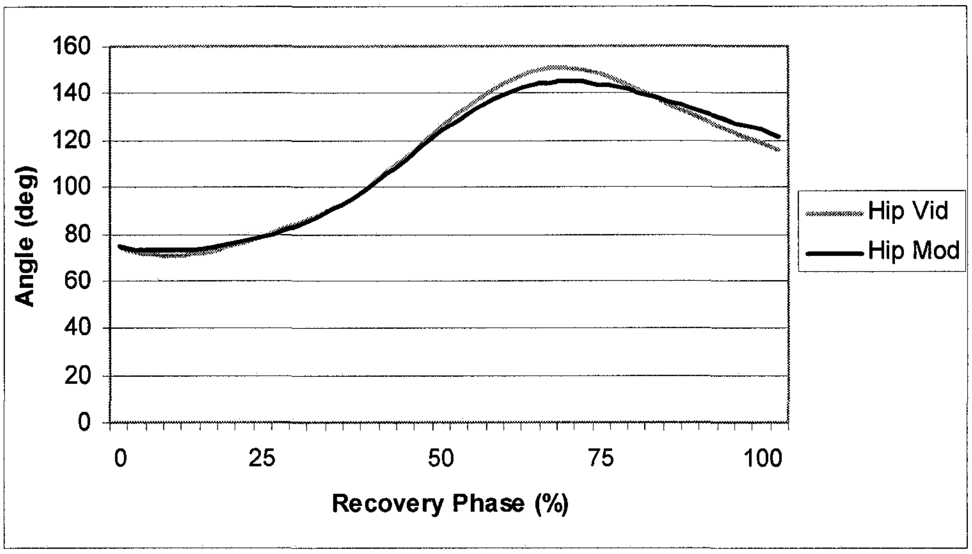
Subject 2 – Hip angular displacement data – video and model



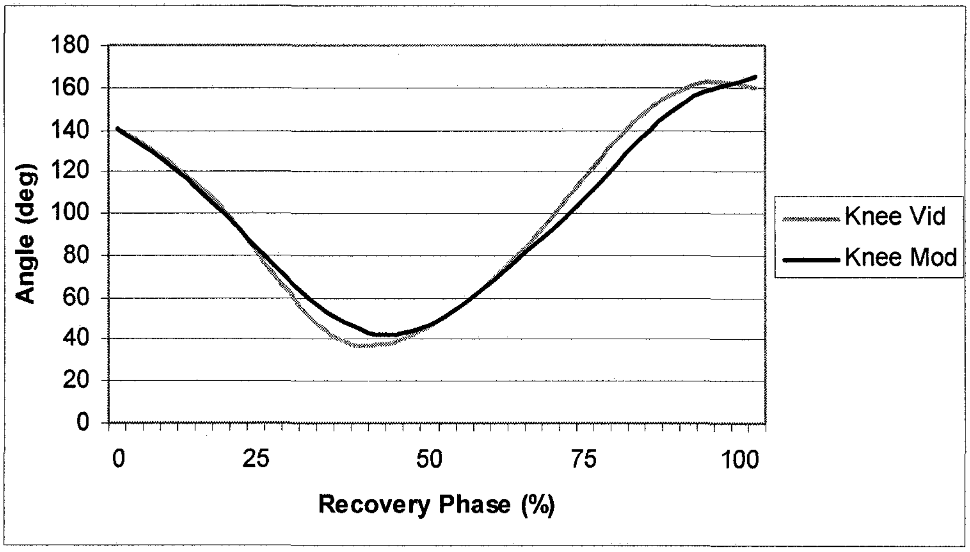
Subject 2 – Knee angular displacement data – video and model



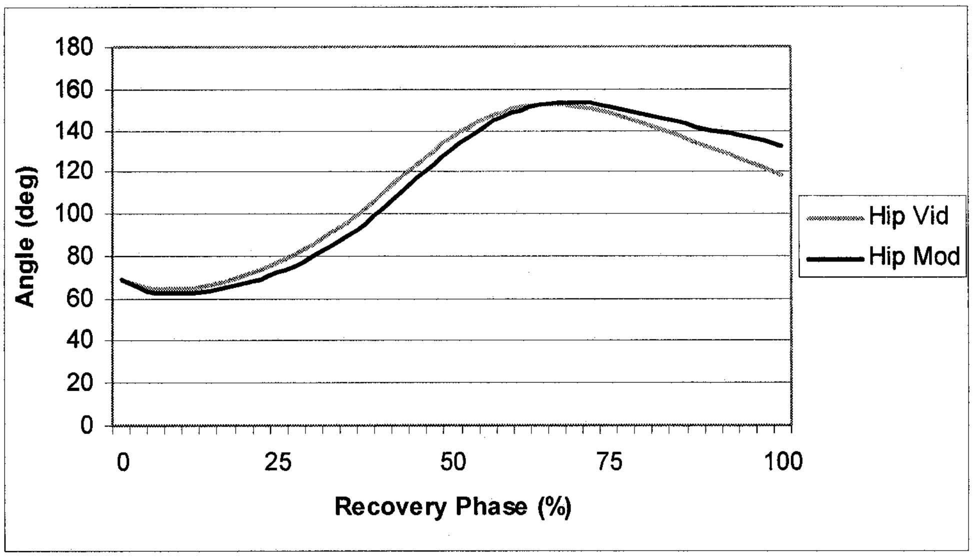
Subject 3 – Hip angular displacement data – video and model



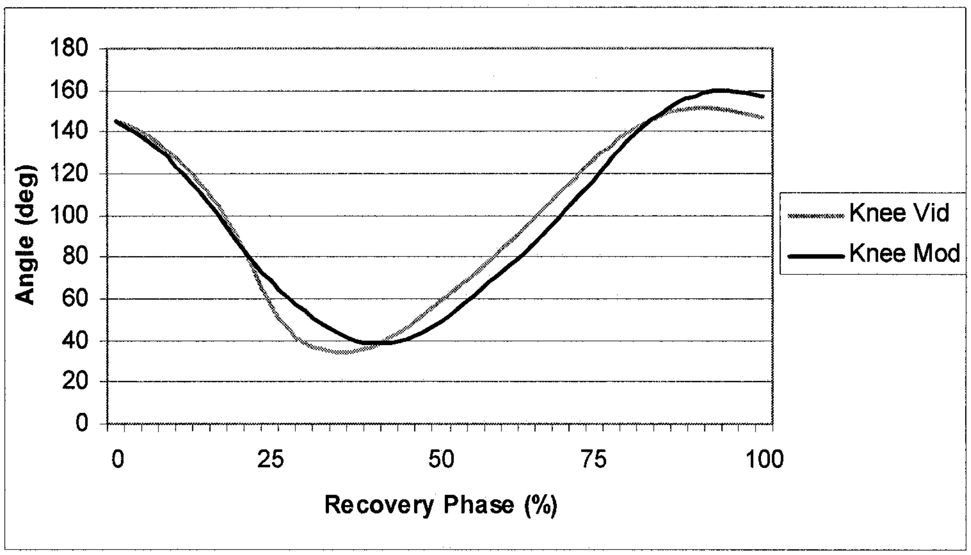
Subject 3 – Knee angular displacement data – video and model



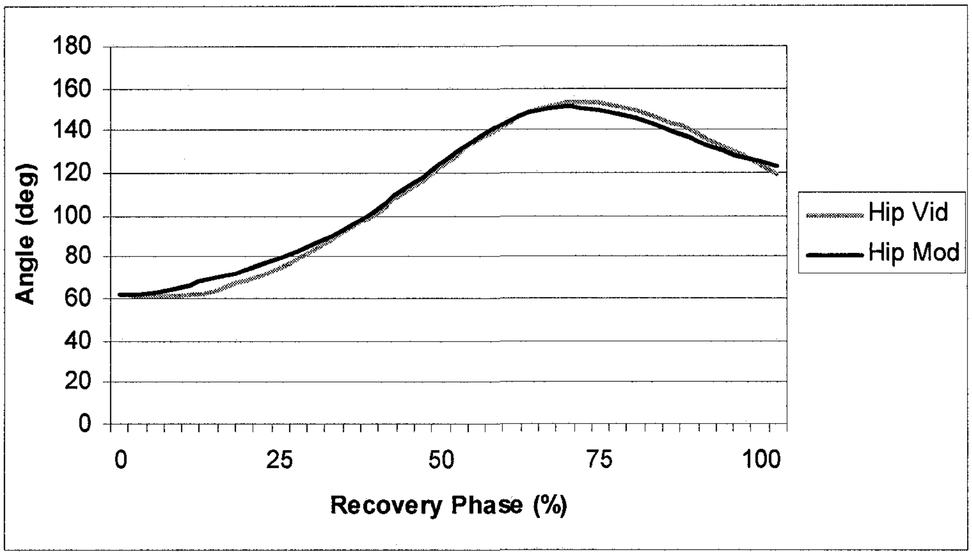
Subject 4 – Hip angular displacement data – video and model



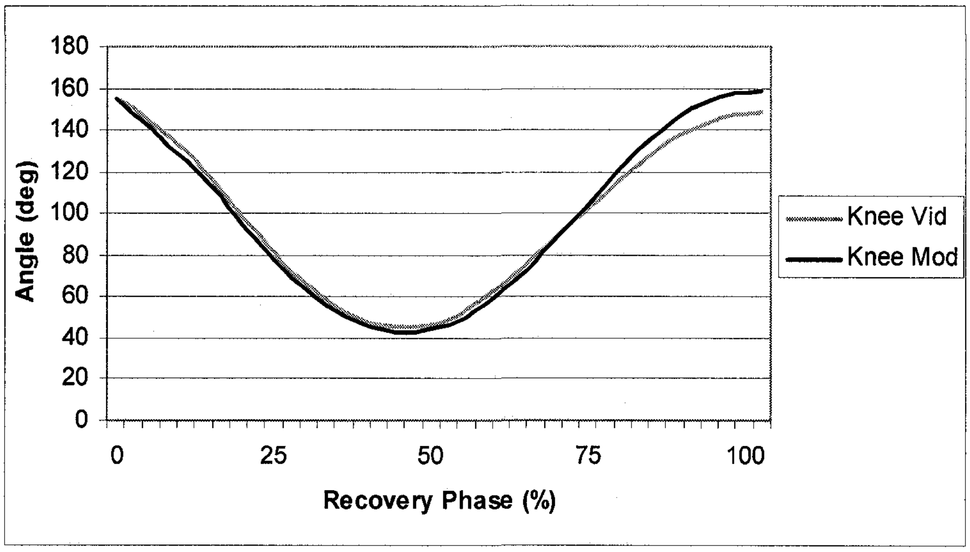
Subject 4 – Knee angular displacement data – video and model



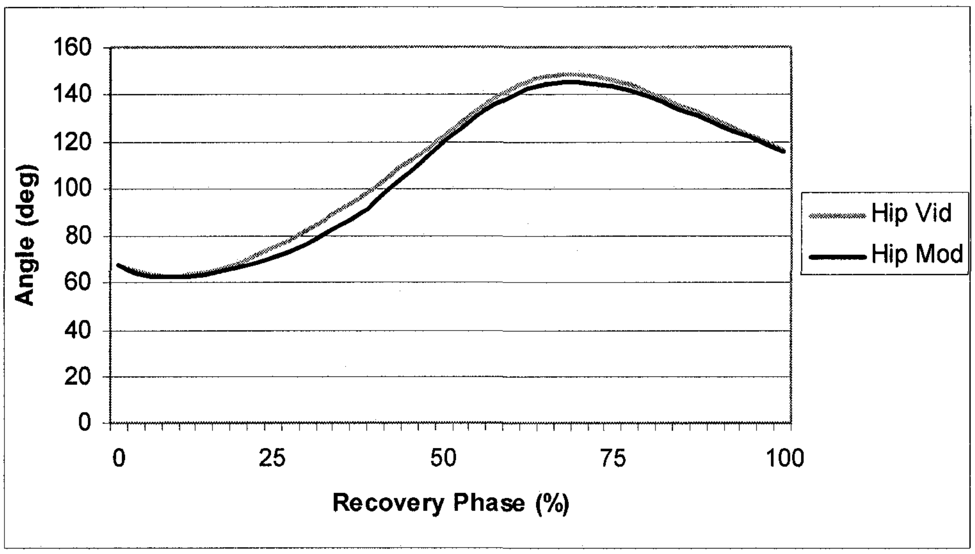
Subject 5 – Hip angular displacement data – video and model



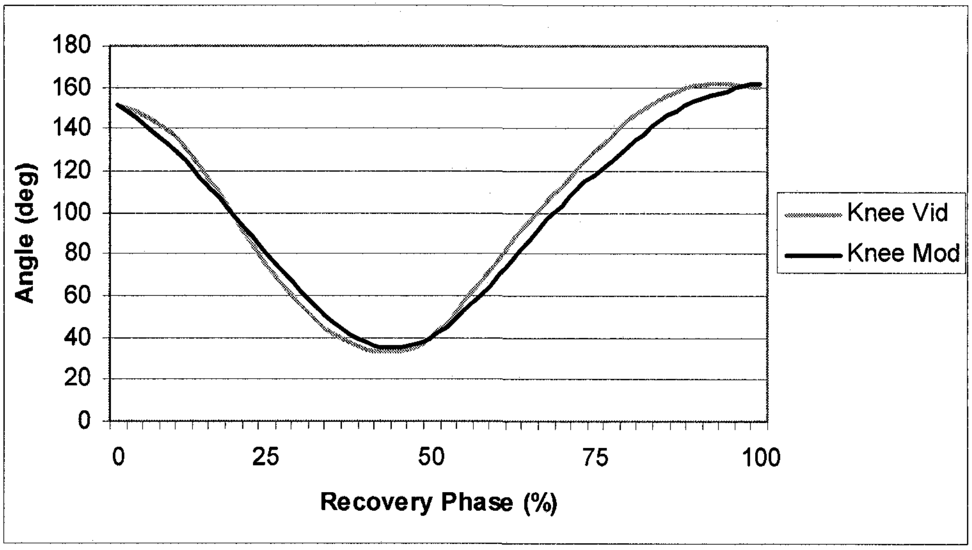
Subject 5 – Knee angular displacement data – video and model



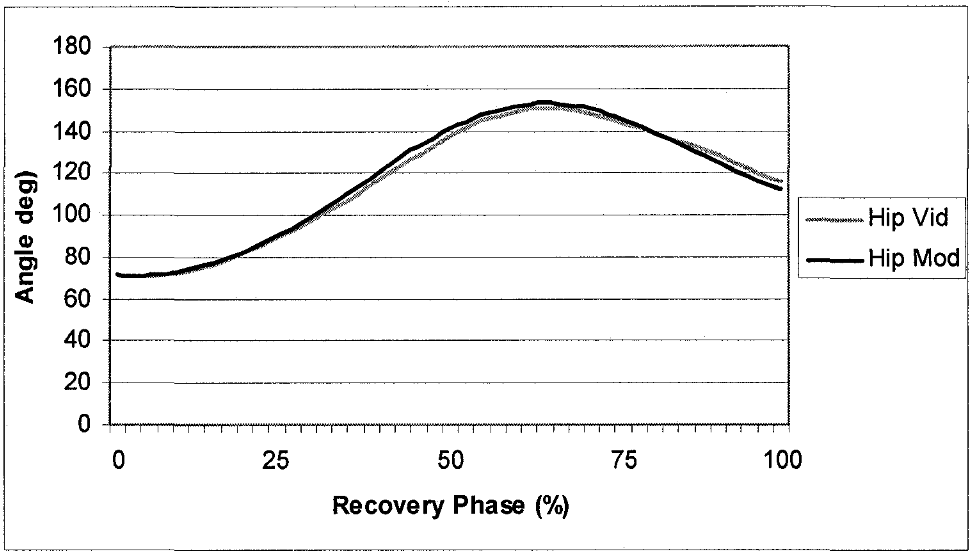
Subject 6 – Hip angular displacement data – video and model



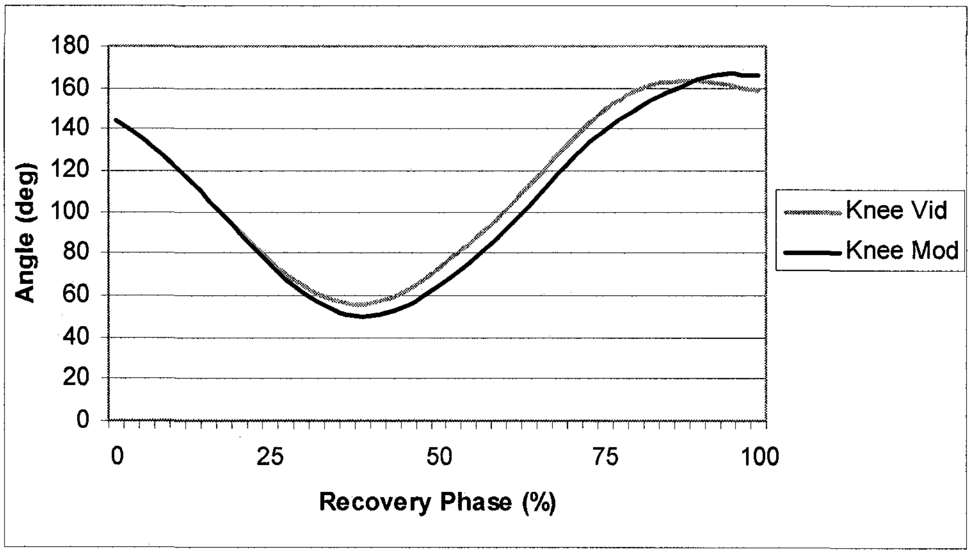
Subject 6 – Knee angular displacement data – video and model



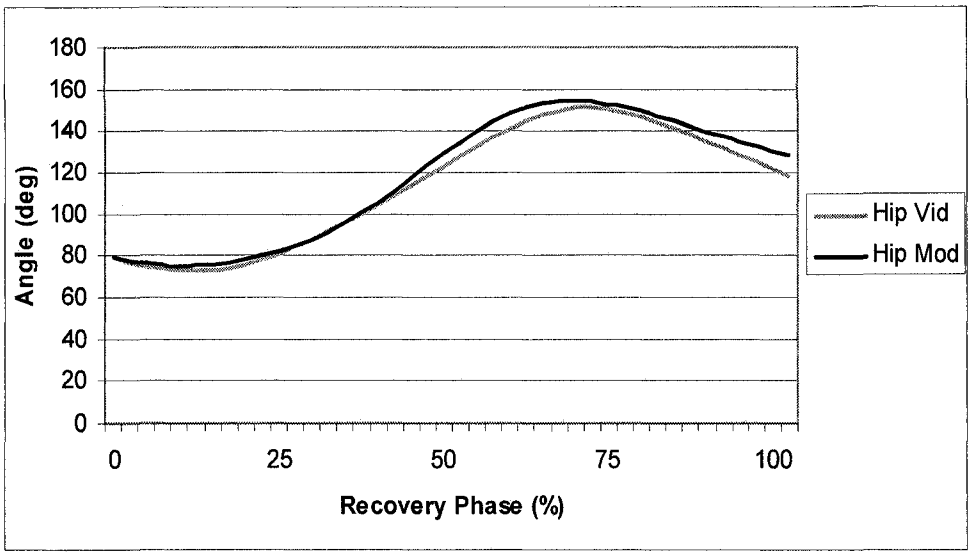
Subject 7 – Hip angular displacement data – video and model



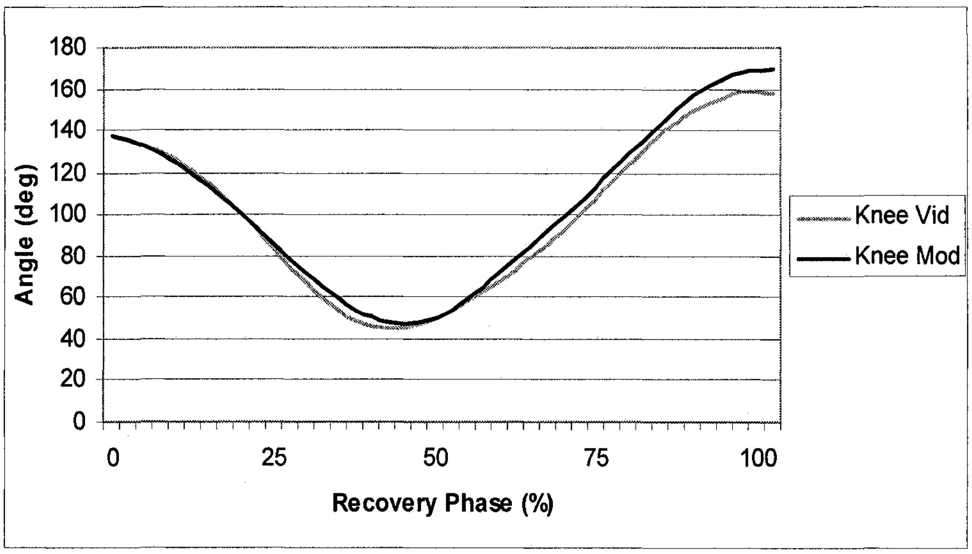
Subject 7 – Knee angular displacement data – video and model



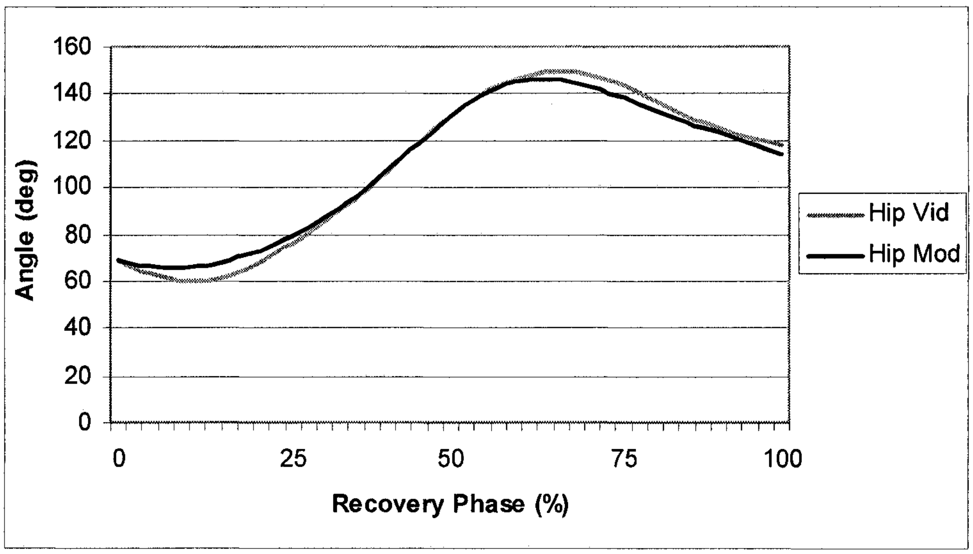
Subject 8 – Hip angular displacement data – video and model



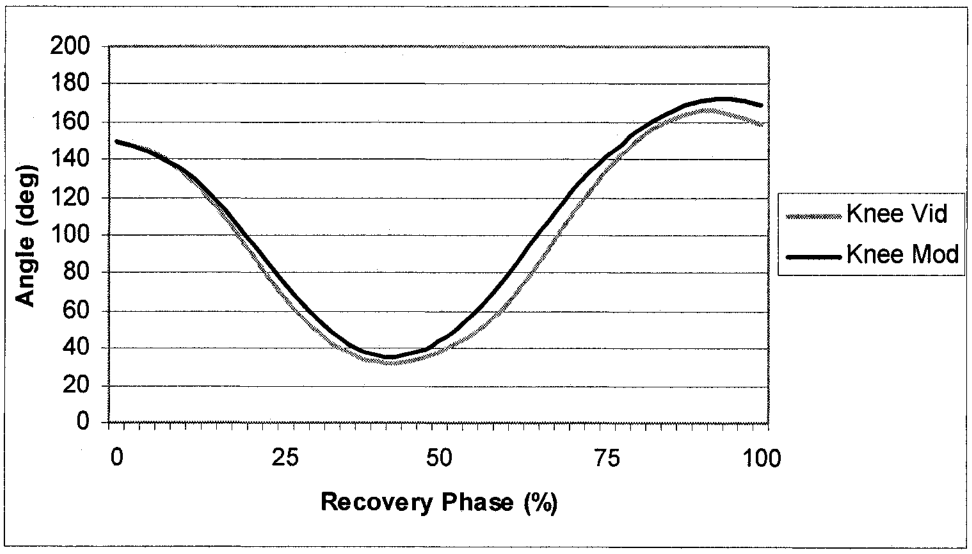
Subject 8 – Knee angular displacement data – video and model



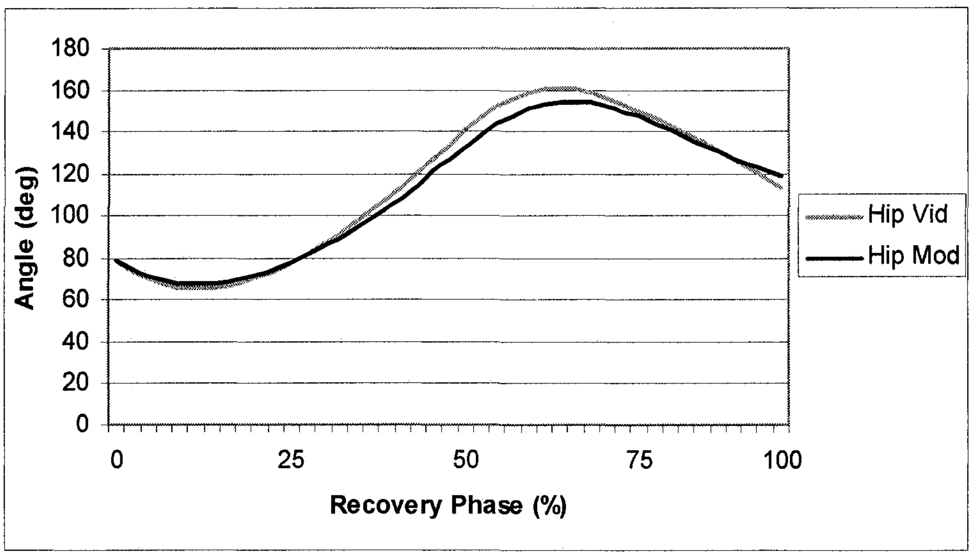
Subject 9 – Hip angular displacement data – video and model



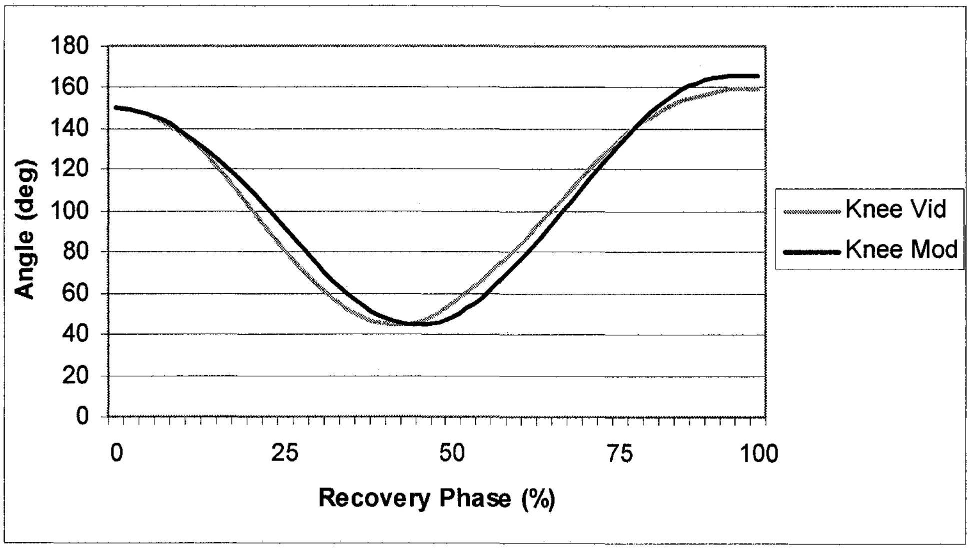
Subject 9 – Knee angular displacement data – video and model



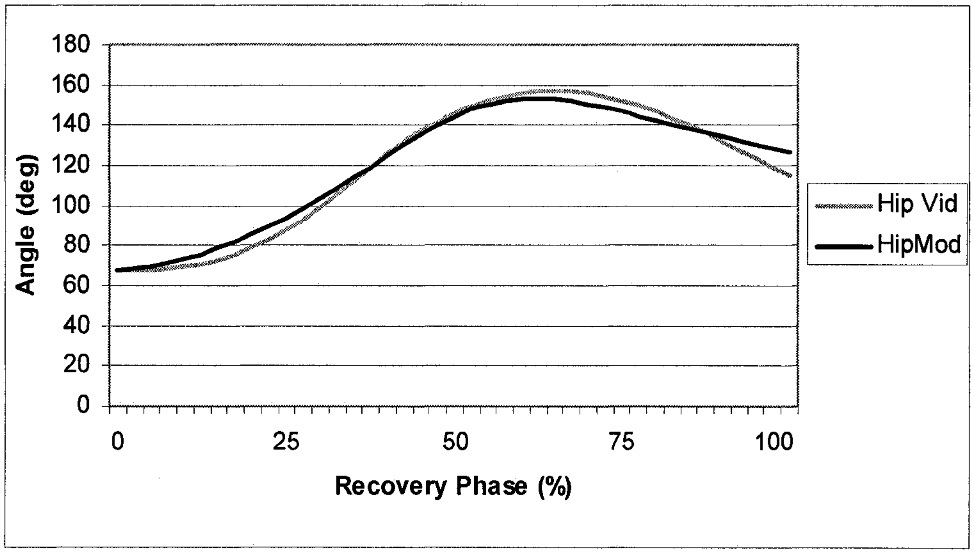
Subject 10 – Hip angular displacement data – video and model



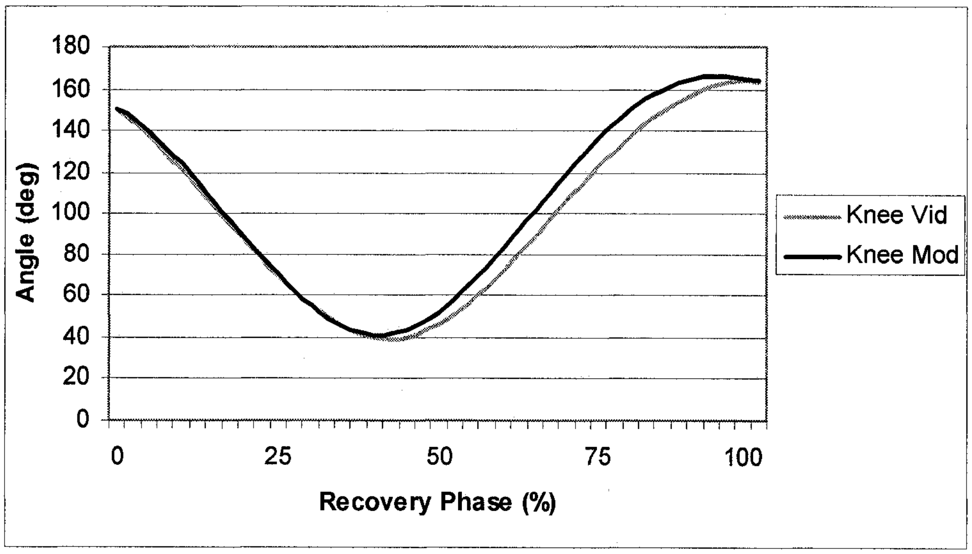
Subject 10 – Knee angular displacement data – video and model



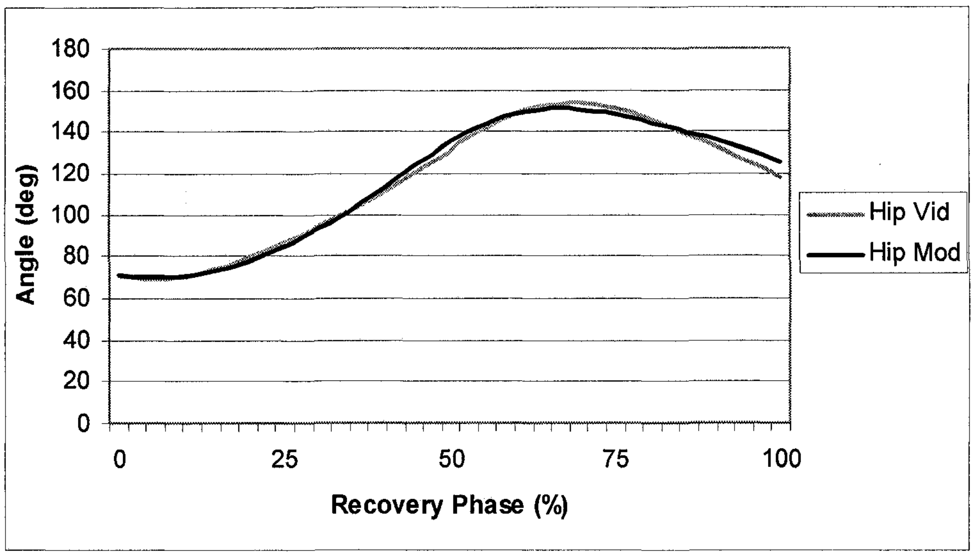
Subject 11 – Hip angular displacement data – video and model



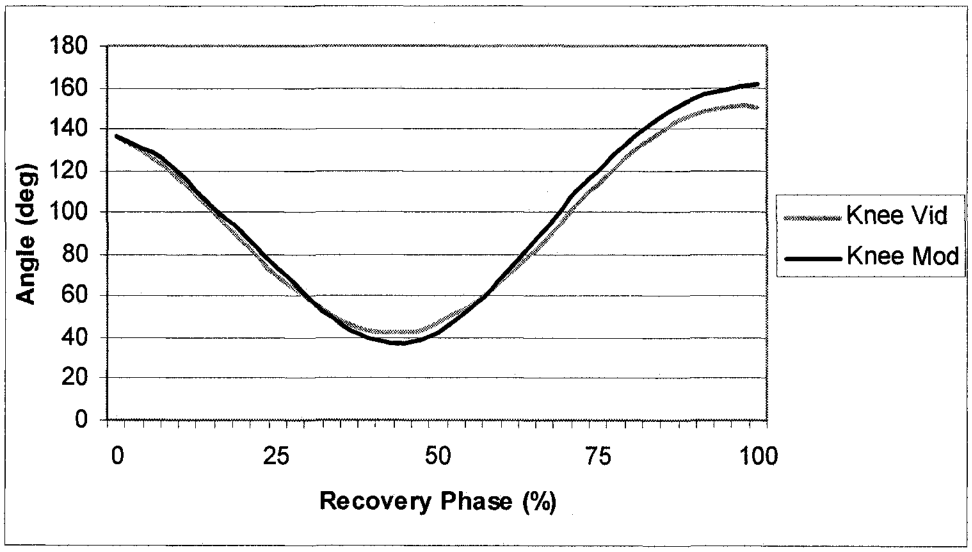
Subject 11 – Knee angular displacement data – video and model



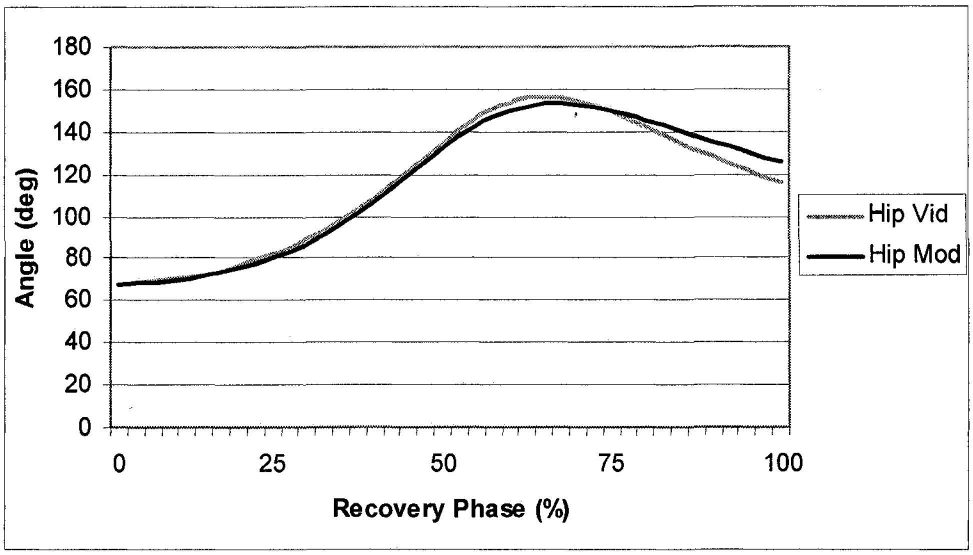
Subject 12 – Hip angular displacement data – video and model



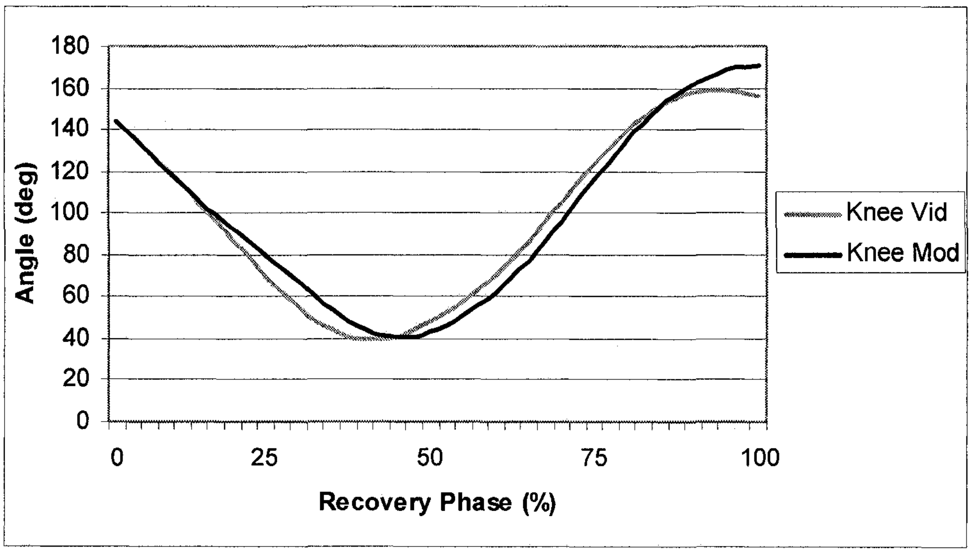
Subject 12 – Knee angular displacement data – video and model



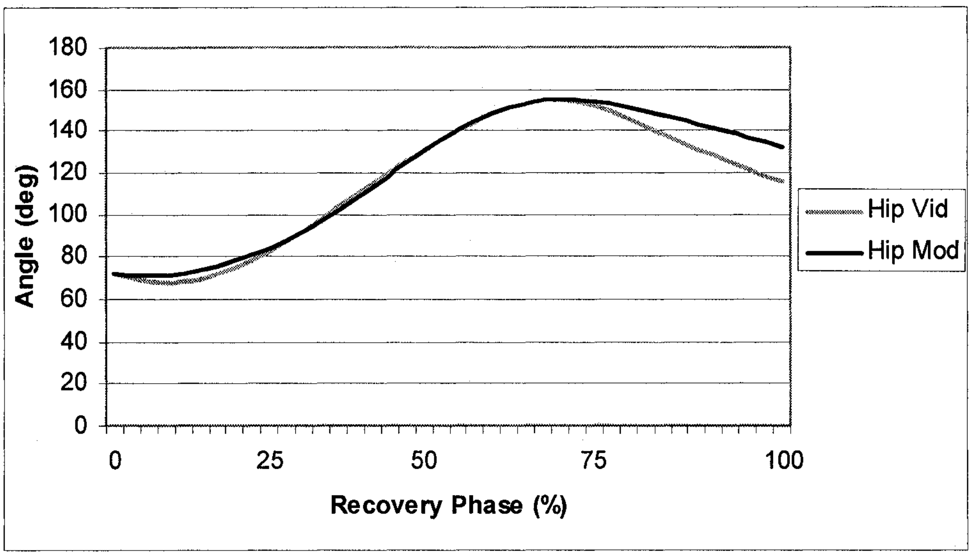
Subject 13 – Hip angular displacement data – video and model



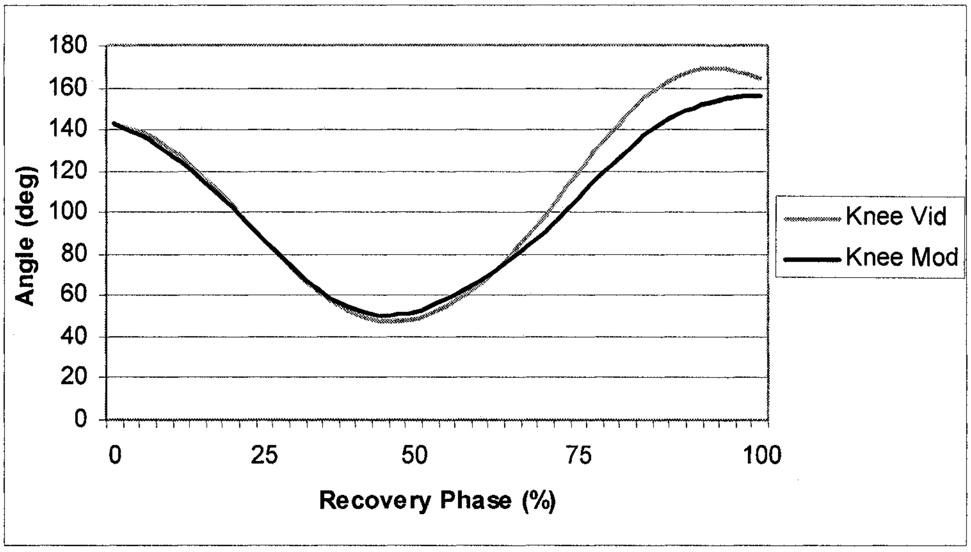
Subject 13 – Knee angular displacement data – video and model



Subject 14 – Hip angular displacement data – video and model



Subject 14 – Knee angular displacement data – video and model



Appendix 3

Review of Literature Specific to Biomechanics: Anthropometrics, Image Analysis, Data Filtering

A3.0 Anthropometrics

Many problems in biomechanics require knowledge of anthropometric data of limb segments about axes through joints, particularly for the quantitative analysis of limb kinetics (Peyton, 1986). Unlike a rigid object where there is a relatively straightforward relationship between the forces producing motion and the translation and rotation of the object, the human body is more realistically represented by a system of linked segments that reposition during movement. The linear acceleration and, consequently, velocity and displacement of a segment are dependent on both the segment mass and the sum of all external forces, including muscle forces, acting on the segment. Angular acceleration, velocity, and displacement are dependent on both the segment principal moments of inertia and the external moments applied to the segment (Winter, 1979). The researcher, then, must know the mass, location of the centre of mass, and the inertia tensor for each body segment. These are the body segment inertia parameters. Without these parameters it is not possible to proceed with simulation or optimization of human movement, using the direct dynamics approach, or to analyze human movement, using the inverse dynamics approach (Reid & Jensen, 1990). Unfortunately there is no universally accepted protocol for the estimation of segmental inertia parameters (Reid & Jensen, 1990).

A3.0.1 Inertia Parameters

Segmental inertia parameters can be calculated using a number of different methods. Each method will be introduced, and the advantages, disadvantages and accuracy will be discussed.

1. Cadaver Data

A number of studies have estimated the segmental inertia parameters on cadavers (e.g. Dempster, 1955; Clauser, McConville & Young, 1969; Chandler, Clauser, McConville, Reynolds & Young, 1975). This method involves dissecting the body into various segments and measuring the mass, mass centre location and moment of inertia of each segment. The mass is found by weighing each segment, the mass centre location is determined by using a balancing plate and the moment of inertia by using a compound pendulum technique (Reid & Jensen, 1990).

The advantage of cadaver studies is that the inertia parameters can be measured directly for each body segment. These parameters can then be used to check the accuracy of the parameter estimates determined from other techniques (Reid & Jensen, 1990).

The primary disadvantages of the cadaver studies are in the sampling and the adequacy of the measurements. Typically, samples are small and do not represent the population under investigation, particularly athletes. There is a lack of consistency in the sectioning of segments and in measurement procedures, making it difficult to compare results of different studies (Reid & Jensen, 1990).

According to Lephart (1984), one cannot be confident in the regression models developed in cadaver studies because of insufficient data. There are two reasons for the lack cadaver data: inadequate numbers of cadavers available for this type of work, and the difficulties in completing the assessment of the inertial properties even when cadavers are available. There are several aspects to this second reason. The equipment used to measure these properties with validity has not been standardized, and before actual cadaver work can be performed it must be extensively tested. In addition, this type of work is tedious and time consuming since the equipment developed to date is delicate and difficult to replace if damaged. Finally, it is often necessary to work within a cold environment since the segments are typically frozen to ensure rigidity and minimize fluid loss, and allowing data to be collected over a period of time.

2. Direct Measurement

Mass centre locations have been measured using reaction boards, segmental volumes by water immersion, and moment of inertia through oscillation techniques (e.g. Drillis, Contini & Bluestein, 1964; Hatze, 1975, Allum & Young, 1976, Peyton, 1986). These methods have a number of disadvantages. It is difficult to judge planes of segmentation and measure segment volumes on living subjects using immersion techniques. In addition, reaction and oscillation techniques are difficult to apply to many segments.

3. Statistical modeling

Dempster (1955) express segment mass and segment mass centre location as percentages of subject mass and segment length and moment of inertia data separately for eight cadavers. More complex types of statistical modeling involve the use of regression equations to relate the segmental inertia parameters to body measurements taken on the subject. Cadaver data have been used to develop regression equations for segmental inertia parameters as functions of anthropometric measurements (Challis & Kerwin, 1992). This technique is most beneficial when average values are required for analysis, or when it is not possible to obtain measurements directly from the participant or athlete. Yeadon and Morlock (1989) developed regression equations from the cadaver data of Chandler et al. (1975) and concluded that non-linear equations were superior to linear equations for providing estimates of segmental moments of inertia, especially when the anthropometric measurements were outside the sample range used to develop the equations. Vaughan, Davis, and O'Connor (1992) devised regression equations to estimate the segment masses and moments of inertia of the lower extremity in their analysis of human gait. In order to use these equations 20 measurements were required, including 18 for the lower extremities, one for the subject's total mass and a measurement of the distance between the anterior superior iliac spines (ASIS).

4. Geometric modeling

Estimating the segmental inertia parameters through the development of geometric mathematical models has been used by Hanavan (1964), Jensen (1978), Hatze (1980), Yeadon (1990b), and Vaughan, Davis, and O'Connor (1992). These models vary in complexity, however, they are all based upon the same principles. A series of geometric solids are used to represent the segments of the body, with the size of the solids estimated from anthropometric measurements and the density of the segments taken from cadaver data. According to Lephart (1984), geometric models can be used to quickly estimate the segmental inertia parameters and the results can easily be displayed on a computer, but the lack of validation of the models should be concern scientists. Hanavan (1964) devised a personalized mathematical model consisting of 15 geometric shapes with segment mass predicted using Barter's (1957) regression equations. The length and diameter of each segment was based on 25 anthropometric measurements. The head was represented by an ellipsoid, upper and lower trunk were right elliptical cylinders, the hands by solid spheres, and the remaining segments of the body were frustums of right circular cones. According to Reid and Jensen (1990), the Hanavan model is easy to use but it oversimplifies the shape of the segments and cannot be considered very accurate.

Jensen (1978) developed a 16 segment model and divided each segment into a series of 2.0 centimetre thick elliptical zones, with density values taken from Dempster (1955). Digitized photographs of the participant were used to estimate the size of each elliptical solid. Jensen reported a maximum error in total body mass for three subjects of different body shapes of 1.8%.

Hatze (1980) developed a mathematical model which included 17 segments and required 242 direct measurements to be taken. Various geometric solids were used to model the segments, and it was not necessary for the segments to be symmetrical. Density values from Clauser et al. (1969) and

Dempster (1955) were used in the model and were varied along the segments. The maximum total body mass error was 0.32% for the four subjects tested.

Yeadon (1990b) developed a model which consisted of eleven segments and required 95 anthropometric measurements to be taken. Errors of approximately 2.3% were measured in total body mass, which were similar to those found by Jensen (1978) but greater than those reported by Hatze (1980). In explaining the sources of error, Yeadon stated that breathing while taking anthropometric measurements makes it difficult to accurately estimate torso parameters. For a 70 kg subject, an extra one litre of air in the lungs would increase the estimate of the total body mass by 1.5%. Yeadon believed that an error of 2.3% was quite reasonable.

For their analysis of human gait, Vaughan et al. (1992) developed a six segment model of the lower extremity, which included the left and right thigh, shank, and foot segments. Regression equations based on the six cadavers studied by Chandler et al. (1975) were developed to predict the body segment parameters for a normal male, which required 20 anthropometric measurements be made.

5. Image Analysis Techniques

Body segment parameters have been estimated by measuring segment lengths from digitized film data in combination with regression equations. This method is essentially the same as for mathematical modeling except the measurements are taken from digitized film data rather than directly from the subject, which is the primary advantage of this technique. A method for estimating segmental inertia parameters from a limited number of segmental length measurements and selected measured perimeters was developed by Yeadon, Challis, and Ng (1994). Using this data, regression equations were then used to calculate the 95 anthropometric measurements required to estimate the segmental inertia parameters using Yeadon's (1990b) mathematical model. Similar results were found between full data set of 95 anthropometric measurements and the limited data set.

Baca (1996) developed a method to obtain estimates of 220 of the 242 measurements required for the geometric model by Hatze (1980) from video by using an automated video imaging system using four different body configurations. Regression equations were used to determine the remaining 22 measurements which could not be obtained from the video. The estimated segmental inertia parameters directly from video were compared with those found from anthropometric measurements, and an average error of less than 5% was found in the total mass for the three subjects analyzed.

Image analysis techniques appear to provide acceptable estimates of segmental inertia parameters, however, it is always preferable that direct measurements are taken from the participants of the study whenever possible.

6. Computer-aided tomography (CT scan) / magnetic resonance imaging (MRI)

CT scanning (Rodrigue and Gagnon, 1983) and magnetic resonance imaging (Mungiole and Martin, 1990) can provide accurate estimates of segmental inertia parameters, however, such techniques are expensive and are not widely available. The Visible Human Project is the development of a complete, anatomically detailed, three-dimensional representation of the normal male and female human bodies through the use of CT scans, MRI and cryosection images. The male was sectioned at one millimetre intervals, the female at one-third of a millimetre intervals. Although the Visible Human Project will likely produce the most accurate measurements of segmental inertial parameters to date, the fact that only one normal male and female have been detailed suggests that the results will likely be inappropriate for most sport biomechanics studies.

7. Radiation Techniques

The use of radiation techniques for estimating segmental inertia parameters is based on the fact that an object will absorb or attenuate high energy rays in proportion to the density, and independent of the composition (Reid & Jensen, 1990). A computer controls the scanning of the object and calculates the results. Baster-Brooks and Jacobs (1975) first used the technique on biological tissue. They compared scanner estimates for a leg of lamb to measurements based on weighing, reaction change, and pendulum techniques and found errors of 1, 2.2, and 4.8% respectively. Zatsiorsky and Seluyanov (1983) further developed the procedure for tests on humans. Segment masses, mass centres, and principal moments of inertia were measured on a large number (n = 100) of adult male subjects with a mean age of 23.8 years, many of which were physical education students. The body was divided into 16 segments, including three segments each for upper extremity, lower extremity, and the torso, and one for the head. Multiple linear regression models were used to predict the inertia parameters, with weight and stature as predictor variables. In a subsequent study (Zatsiorski & Seluyanov, 1985), these regressions were supplemented by a further set in which segment anthropometric measures were used as predictor variables with segment-specific variables used to improve the prediction accuracy. Reid and Jensen (1990) felt the results from these studies provided a good representation of adult males; however, there is a limitation to the applicability of this data to biomechanics research. This is because bony landmarks were used as reference points to locate segmental endpoints and to define segment lengths, rather than using joint centres (DeLeva, 1996). Some of these bony landmarks are relatively distant from the joint centres which they are meant to define, which results in inaccurate anthropometric values. DeLeva (1996) made adjustments to these segmental endpoints to match them up with the joint centre locations which are typically used in biomechanics research, and made the appropriate corrections to the segmental lengths, centre of mass locations, and radii of gyration.

A3.1 Image Analysis

When the kinematic and kinetic measurement of segmental motion is required for the analysis of motion and for the implementation of a simulation model, a problem that arises is how to measure the variables as they cannot be measured directly. The typical procedure involves recording the positions of various body landmarks at discrete intervals in time, and other kinematic variables are obtained by numerical differentiation (Wood, 1982). The most common way to obtain time histories of these landmarks during athletic movements is to digitize cinefilm or video recordings of the performance (Yeadon & Challis, 1994). An alternative method uses automated motion analysis, which is becoming more popular as the technology improves (e.g. PCReflex which uses reflective markers to track the motion of the body parts). An automated system has the advantage of not having to manually digitize body landmarks frame by frame, but is subject to tracking discontinuities associated with markers being obscured by moving body parts. In addition, automated systems are often impractical in a field setting. The following sections will present the literature relevant to the collection of time histories of various body landmarks from recorded images, including a discussion of filming techniques, 2D and 3D techniques and reconstruction, and lens distortion.

A3.1.1 Filming Techniques

The most frequently used method of collecting kinematic data involves the use of an imaging or motion-capture system to record the movement of a subject, followed by a digitizing process from which the kinematic variables are calculated to describe the segmental or joint movements (Robertson & Caldwell, 2004). The most common imaging systems use video or digital video cameras (e.g., APAS or Peak). Motion is recorded using ambient light or light reflected by markers located on anatomical landmarks on the body. When used in controlled, laboratory settings, external lights located near the cameras amplify the brightness of the markers so they stand out compared to the background, clothing, or skin. Some systems use active infrared light-emitting diodes or IREDs (e.g., Optotrack), with others using reflective infrared light (e.g., Vicon).

The active marker systems require a control unit that pulses the individual IREDs in a specific sequence for marker identification (Robertson & Caldwell, 2004).

Before the introduction of video systems, the method of choice for recording motion was 16 mm cinefilm. At one time cinefilm was considered to be a better quality recording medium to use than video, although now both methods are considered accurate from a practical standpoint (Kennedy et al., 1992). According to Robertson and Caldwell (2004), cinefilm has a number of advantages over video, including a wider range of camera frame rates and shutter speeds, and finer image resolution. Cinefilm analysis, however, requires much more time to manually digitize the coordinates, often requiring hours to digitize a few seconds of film. When combined with the delays associated with film processing, fast data turn around times are not feasible. Also, there is no way to view the film during or immediately after the testing session so errors are not discovered until long after. Video systems allow real-time viewing of each trial and immediate replay to check the veracity of the recorded images.

The use of high speed cinematography has certain advantages over video filming, however, Robertson and Sprigings (1987) and Abraham (1987) stated that the cost of high speed cinematography has made the use of video recording for movement analysis a popular option for many researchers. Abraham (1987) also stated that for moderate speed movements, the more affordable video system provided reasonable image resolution, freeze-frame analysis, and single frame advance. Kennedy et al. (1989) and Robertson and Sprigings (1987) felt that video analysis was easier to use and did not require processing time. The images are immediately available, thereby allowing the investigator to observe the image quality during the recording session (Angulo & Dapena, 1992). Another advantage of the modern imaging systems is the capability of automated digitizing, which allows for the positional data from multiple joint markers during an entire movement sequence to be quickly calculated and displayed (Robertson & Caldwell, 2004). However, when the testing takes place in an uncontrolled environment such as a national or international competition, it is not possible to apply markers to the subjects. In these situations the digitizing must be completed manually.

According to Angulo and Dapena (1992), there are a number of limitations of the video recording technique, including a limited number of frames that can be filmed per second, the quality of the video image, and the accuracy of the coordinate values due to limitations in pixel size. Cinefilm cameras are capable of filming at 500 frames per second or more, while conventional video or digital video cameras record at frames per second (or 60 fields per second). However, with the development of digital technology, digital cameras are now available which record at high frame rates as well. Other cameras, such as the JVC GVL-9800 allow for increased frame rates by changing the number of fields which are recorded on one video frame, thereby allowing for recording speeds of up to 240 fields per second.

One of the most important factors when comparing cinefilm to video is the accuracy of the coordinate values. This is influenced by the resolution and quality of the video image; which is limited by the size of the pixels used on the monitor (Angulo & Dapena, 1992). Kennedy et al. (1989) compared the accuracy of predicting the points in the x, y, and z planes for the two filming methods. They found the average error of the coordinates of the points of their 2 metre control object to be 4.8 mm for film, and 5.8 mm for video. The researchers stated that the 1 mm difference was not large enough to consider cinefilm to be more accurate in terms of point production, even though the two methods were found to be significantly different ($p < 0.05$).

Angulo and Dapena (1992) found that when using a large field of view (8 meters) there was a greater error in accuracy for video, with a resultant error of 10 mm recorded for the reconstructed coordinates for the control object. In comparison, resultant errors of 4 and 5 mm were found for the large and small film images, respectively. For the external landmarks in the "xy" plane, the resultant error for the video technique was larger (39 mm) than the larger (29 mm) and smaller (28 mm) film image techniques. Although the accuracy of the video analysis technique was affected by the larger view, the authors noted that within the volume of the control object, the video technique was accurate enough for most applications.

A3.1.2 2D and 3D Techniques

When reconstructing digitized landmarks using a two-dimensional technique, it is assumed that all the points are located on a single plane. Digitized points that are actually located outside the plane of reconstruction are projected on to the plane of reconstruction (Walton, 1981), and movements occurring outside the plane are ignored. This may limit the accuracy of the findings (Yeadon and Challis, 1994). In order to justify using a 2D analysis, it is suggested that a comparison of 2D and 3D reconstruction techniques for a given movement should be performed (Yeadon & Challis, 1994). For the chosen activities in the present study (sprinting), previous studies have predominantly used 2D reconstruction techniques (e.g. Mann & Herman, 1985) as a sagittal plane analysis can identify the majority of the important details about running (Robertson & Caldwell, 2004). The advantage of 2D analysis is that only a single cine/video camera is required to record the activity from which the body landmarks can be digitized and reconstructed. In contrast, a 3D analysis requires at least 2 cameras and is far more complex (Yeadon and Challis, 1994).

A3.1.3 2D and 3D Reconstruction

The most frequently used method for reconstructing coordinates of digitized body landmarks recorded using two or more cameras with fixed orientation is the Direct Linear Transformation (DLT) technique (Abdel-Aziz & Karara, 1971). The DLT technique allows for the cameras to be located arbitrarily, however, it also requires that the calibration control points are evenly distribution throughout the control space (Abdel-Aziz & Karara, 1971). The DLT reconstruction relates the three-dimensional object space and the two-dimensional image plane through the camera using two equations with 11 parameters:

$$x + \delta x + \Delta x = \frac{L_1 X + L_2 Y + L_3 Z + L_4}{L_9 X + L_{10} Y + L_{11} Z + 1}$$

$$y + \delta y + \Delta y = \frac{L_5 X + L_6 Y + L_7 Z + L_8}{L_9 X + L_{10} Y + L_{11} Z + 1}$$

where X,Y,Z are the three space coordinates of a point in the object space, x,y are the two coordinates of the same point mapped into the image coordinate system, δx and δy are nonlinear systematic errors, Δx and Δy are random errors and the L_i, i = 1, ..., 11, are referred to as the 11 DLT parameters. Of the 11 geometrical parameters, 6 define the location and orientation of the camera and the other 5 define the characteristics of the digitizing measurement system.

The DLT method involves the calibration of each camera view by calculating the values of the 11 parameters from the digitized coordinates of the control points which are in know locations. A minimum of six control points are required to solve for the 11 parameters, however, more points are often used and recommended (Wood & Marshall, 1986).

The DLT method may also be applied to two-dimensional analysis. For 2D DLT, the two equations are reduced to:

$$x + \delta x + \Delta x = \frac{L_1X + L_2Y + L_3}{L_7X + L_8Y + 1}$$

$$y + \delta y + \Delta y = \frac{L_4X + L_5Y + L_6}{L_7X + L_8Y + 1}$$

since z is always zero. At least one camera and four control points are required for 2D DLT analysis.

The accuracy of the DLT method for 3D reconstruction has been examined by a number of researchers. Wood and Marshall (1986) and Chen, Armstrong, and Raftopolous (1994) found that the number of control points used in the calibration of each camera and their distribution in the control space affected the reconstruction accuracy. Chen et al. (1994) found that the accuracy improved as the number of control points increased from 8 to 24 and demonstrated that the 'best accuracy' was when the control points were evenly distributed throughout the control space. In addition, both Wood and Marshall (1986) and Chen et al. (1994) found that reconstruction errors increased significantly outside the calibrated volume. Because these points lay outside the scaled control area, their positions must be interpolated by the computer since there are no surrounding values with which they can be compared. Therefore, this causes an increase in the likelihood of an error being introduced into the data collection. In addition, Wood and Marshall (1986) recommended a camera setup where the optical axes of the cameras were perpendicular, although the cameras can be set up at any angle to one another (Shapiro, 1978).

Once the 11 DLT parameters for each camera have been calculated, they can then be used to reconstruct the locations of the digitized body landmarks, giving the location of the digitized points from each camera view. For a 3D analysis with two cameras, a least squares technique can then be used to calculate the 3D coordinates of each digitized point in space, as long as a minimum of 6 control points are used (Shapiro, 1978).

Hatze (1988) made a modification to the DLT procedure (MDLT) by constraining one of DLT parameters using a constraint equation. By doing so, the number of parameters was reduced to from 11 to 10. Hatze found a large increase in reconstruction accuracy from 5 mm to 1 mm.

Ideally, all cameras used during analysis will record each marker at every instant in time. Unfortunately, in real situations this is not the case and markers are frequently hidden from view. In video analysis, these markers are lost until they are once again visible to the camera. The position of a marker at a given point in time is calculated using only those cameras that "see" the marker, which may have an influence on the number of cameras used and their placement during data collection. In general, increasing the number of cameras reduces the potential noise in the data (Nigg & Cole, 1994). A greater number of cameras, however, does not guarantee there will be an increase in the accuracy of the determined marker positions.

A3.1.4 Lens Distortion

With the DLT technique it is assumed that there is an "ideal camera-digitizer lens system" in which the point in space, the image point, and the centre of the lens all lie on a straight line, but this is not always the case. The effects of non-linear lens distortion in the DLT procedure have accounted for by various researchers in which extra parameters were added to the two DLT equations (Wood & Marshall, 1986; Hatze, 1988). Wood and Marshall (1986) added one extra parameter to the 11 parameter DLT to provide a partial correction for the non-linear symmetrical lens distortion. The accuracy in the reconstruction was unchanged, and may have been due to the high quality of the lenses or the small number of control points used. Hatze (1988) added lens correction to his MDLT technique to account for both radial and asymmetrical lens distortion (up to 5 extra parameters). Only a slight improvement in reconstruction accuracy was found.

A3.2 Data Filtering

Through the process of obtaining time histories of various body landmarks, it is inevitable that a certain amount of error is introduced to the signal. The error associated with positional data is referred to as noise, which is the component of the final signal that is not due to the process itself (Winter, 1990). Sources of noise in biomechanics research include: movement of the cameras, movement of the body markers, human error in digitizing, and limitation in the precision of the digitizing process. This error is usually at a higher frequency than the signal itself, which is located at the lower end of the frequency spectrum (Winter, 1990). Velocity calculations require the positional data to be differentiated once, and the accelerations require double differentiation of the positional data. Because differentiation preferentially amplifies the higher frequency component of the signal (Wood, 1982), various methods have been devised to help reduce the noise by selectively reconstructing the signal up to a frequency that does not include this unwanted component. These methods, generally referred to as data smoothing or data filtering techniques, are very important in biomechanical analysis, particularly when investigating subtle differences among elite level performers. Wood (1982) suggests using 1) Spline curve fitting, 2) Fourier smoothing or 3) Digital filtering. These routines provide an adequate description of the displacement-time data, while at the same time minimizing measurement error.

A3.2.1 Spline Curve Fitting

Spline functions piece together a number of different low degree polynomials, with the junction point of the different functions known as knots. Cubic (3^{rd} order) and quintic (5^{th} order) splines are most frequently used. There are three decisions that need to be made when curve fitting using splines: the degree of spline, how accurate the spline is to be, and the number of knots to be used. The general rule of thumb when using splines is

> 1. there should be as few knots as possible, ensuring that there are at least four or five points per interval; 2. there should not be more than one extremum point ... or one inflection point per interval; 3. extremum points should be centred in the interval; 4. inflection points should be close to the knots (Wood, 1982, p.327).

The fact that the final smoothed data is represented by a series of equations means that the line can adapt quickly to rapid changes in direction (Wood, 1982). Quintic splines are preferred over cubic splines where second or higher order derivatives are required (Philips & Roberts, 1983) as a cubic spline has a piecewise linear second derivative with restrictions placed upon the endpoint conditions (Challis & Kerwin, 1988; McLaughlin, Dillman & Lardner, 1977). Fitting a spline to the data allows derivatives to be obtained without any extra data processing. The advantage of using splines is that they are very flexible in allowing data that is not equispaced to be fitted, although problems at the endpoints may occur and a large number of coefficients are required to define a spline (Wood, 1982). While Wood (1982) and Challis and Kerwin (1988) believed splines are an acceptable method of data smoothing, there are a large number of variables that must be taken into consideration when using this method. Wood (1982) also stated that the use of this method requires that there be sufficient data points and that the accuracy of the data is known.

A3.2.2 Fourier Smoothing

Fourier smoothing involves transforming the data to the frequency domain, removing the unwanted frequency coefficients, and then reconstructing the original data without the noise (Derrick, 2004). Fourier smoothing uses a series of sine and cosine curves of increasing frequency to fit the curve (Wood, 1982). A cut-off frequency is then selected with the data reconstructed up to that cut-off frequency. The difficulty with this technique is in the selection of the cut-off frequency. Hatze (1981c) fitted a Fourier series to angular displacement data from Pezzack, Norman, and Winter (1977), and achieved satisfactory approximations of the displacement and acceleration data. In addition, Hatze showed that the Fourier smoothing technique could still produce good acceleration estimates when an extra 5% error was introduced to the original data. Hatze, though, assumed that the second derivatives at the endpoints were zero, which limits the applicability of his method because this is not always the case. When the derivatives at the endpoints are not zero, extra data points can be added to the endpoint so that within the required range the derivatives are not zero (Vint & Hinrichs, 1996; Philips & Roberts, 1983). Overall, the Fourier smoothing appears to be a good alternative for the estimation of high order derivatives, and has the advantage that relatively few coefficients are required to define a series (Wood, 1982).

A3.2.3 Digital Filters

Digital filtering is designed to read data from equally spaced time intervals, reduce the noise, and produce data that closely resembles the original data (Wood, 1982). The format of a digital filter, which processes the data in the time domain, is as follows (Winter, 1979):

$$X^1(nT) = a_o X(nT) + a_1 X(nT - T) + a_2 X(nT - 2T) + b_1 X^1(nT - T) + b_2 X^1(nT-2T)$$

where: X^1 refers to the filtered output coordinates
X refers to the unfiltered coordinate data
nT is the nth sample frame
$(nT - T)$ is the $(n - 1)$th sample frame
$(nT - 2T)$ is the $(n - 2)$th sample frame
$a_o, \ldots, b_2$, etc. are the filter coefficients

These filter coefficients are constraints that depend on the sampling frequency, the cut-off frequency, and the type and order of the filter. The filtered output $X^1(nT)$, is a weighted version of the immediate and past raw data plus a weighted contribution of past filtered output. The order of the filter determines the sharpness of the cut-off, in which the higher the order the sharper the cut-off. However, higher order filters also require a larger number of coefficients. In addition to the attenuation of the signal a phase shift is also seen in the output data, which results in a phase distortion. In order to cancel out this phase lag the filtered data is filtered one more time, but this time in the opposite direction. This recursive procedure results in a net phase shift of zero. In addition, the cut-off filter will be twice as sharp as for single filtering (Winter, 1979).

Butterworth filters, a type of digital filter, are often used to remove high frequency noise from digital data with a cut-off frequency of 4 Hz to 8 Hz for human movements (e.g., Bruggemann, 1987). However, there are a number of disadvantages in using a digital filter. One problem associated with this method is that the investigator must decide which frequency must be used to smooth the data (Wood, 1982). Another problem associated with digital filtering is the slight distortion that occurs where the signal and noise overlap (Winter, 1990). Thirdly, the data must be equispaced (Wood, 1982).

Wood (1982) found that digital filtering, Fourier smoothing, and spline smoothing produce valid results for motion analysis. Therefore, the method of filtering to be used depends on the data to be smoothed, the investigator's preferences, and the availability of the program routine. Challis and Kerwin (1988) tested quintic splines, truncated Fourier series and Butterworth filters using a series of test functions and found that quintic splines were superior in the determination of second derivatives from noisy data. Vint and Hinrichs (1996) also tested cubic splines, quintic splines, Fourier series, and Butterworth filters on modified raw angular displacement data from Pezzack et al., (1977) and also found quintic splines produced the most accurate acceleration estimates. The problem, however, is that the data used in these studies is very different from the displacement, velocity, and acceleration curves seen in the lower extremities during sprinting. The results obtained when comparing filtering techniques may be applicable to the specific data set that was tested, but different results may have been found if data from other sources was used. Within the running and sprinting biomechanics literature the most popular data filtering method appears to be the digital filter (e.g., Hunter, Marshall, & McNair, 2004; Kivi, Maraj, & Gervais, 2002a; Stefanyshyn & Nigg, 1998; Jacobs, Bobbert, & van Ingen

Schenau, 1993; Bobbert, Yeadon, & Nigg, 1992; Chengzhi & Zongcheng, 1987; Yokoi, Shibukawa, Ae, & Hashihara, 1987; Hinrichs, Cavanagh, & Williams, 1983).