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A Three-Dimensional Analysis of Subtalar and Knee Joint Coupling During Running Over Obstacles

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A THREE-DIMENSIONAL ANALYSIS OF SUBTALAR AND KNEE JOINT COUPLING DURING RUNNING OVER OBSTACLES

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THESIS ACCEPTANCE

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A THREE-DIMENSIONAL ANALYSIS OF SUBTALAR AND KNEE JOINT COUPLING DURING RUNNING OVER OBSTACLES

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Running has been associated with a number of lower extremity overuse injuries. Attention has been given to biomechanical factors, specifically, excessive pronation and excessive tibial rotation. It has been suggested that excessive tibial rotation is due to excessive foot pronation transferred through a coupling mechanism. The ankle and knee are mechanically linked via the tibia and excessive tibial internal rotation may delay tibial external rotation as the knee begins to extend. Increased impact forces have also been implicated as a cause of running injuries, although little is known about this possible relationship. Obstacle heights have been used previously to produce increases in impact forces. Investigation of biomechanical factors has traditionally been two-dimensional. However, recent literature has shown limitation to two-dimensional analysis. Therefore, the purpose of this study was to investigate the coupling mechanism between the subtalar and knee joints during running over obstacles of varying heights using a three-dimensional analysis.

Ten, heel strike subjects ran at a self-selected pace under a no obstacle condition and four obstacle conditions (5, 7.5, 10, & 12.5% of standing height) on day 1 and underwent an orthopedic exam on day 2. The obstacle was placed directly before a force platform (960 Hz). Videography was collected using two high-speed cameras (240 Hz). Seven reflective markers were placed on the right limb to identify a three-dimensional 3-segment model.

Increasing obstacle height resulted in increased impact forces. This allowed examination of the coupling mechanism over a spectrum of various impact force magnitudes. The pronation curve transitioned from a unimodal to a bimodal configuration and the bimodal tibial rotation curve experienced increases in the bimodal characteristics. Increasing impact forces resulted in increases of maximum pronation and maximum tibial internal rotation as well as decreasing the time to reach maximum knee flexion. However, the times to maximum pronation or maximum tibial rotation remained unaffected. This resulted in increases in the time differences between maximum pronation and maximum knee flexion. Therefore, the tibia may have been put under abnormal torsional stresses that were augmented with increasing impact forces as the proximal end began external rotation due to earlier knee flexion and the distal end maintained internal rotation due to unchanging pronation times. Future studies will focus in the measurement of these forces, as well as to justify these phenomena with additional perturbations.

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CHAPTER 1: INTRODUCTION

Physical activity is one of the primary modes for achieving the goals and objectives of Healthy People 2010 (U.S. Department of Health and Human Services, 2000). One of these goals is to increase the proportion of adults who engage in physical activity, promoting the development and maintenance of cardiorespiratory fitness, for three or more days per week for 20 or more minutes per occasion. It is also noted that for even greater health benefits, vigorous physical activity is necessary. For most persons, the greatest opportunity for physical activity is associated with recreation and leisure time, because few occupations today provide sufficient vigorous or moderate physical activity to produce health benefits. To achieve the cardiorespiratory goal and satisfy the ability to perform physical activity during recreation and leisure time, a popular mode performed is running.

Novacheck (1998b) has reported that approximately 30 million Americans run for recreation or competition. Fields (1994) estimated that running has 50 million devotees who choose this activity for their primary method of fitness. However, running has been associated with a number of injuries and the estimated yearly incidence of injury may be anywhere from 25-65% (van Mechelen, 1992; Wen, Puffer, & Schmalzried, 1997; Brill & Macera, 1995; Novacheck, 1998a; Walker, 98). Furthermore, approximately 70-80% of these injuries occur in the lower extremities and 50-80% are of the overuse type (Brill & Macera, 1995). The majority of these lower extremity overuse injuries occur at the knee. Knee injuries comprise approximately 30-50% of all running injuries (Fredericson, 1996). Anterior knee pain, an overuse injury often called "runner's knee", is the most common knee complaint and accounts for 29% of all running injuries (Novacheck, 1998b; Browning & Donley, 2000). Therefore, to prevent and treat such injuries, it is essential to understand the mechanisms that predispose and lead to injury.

Much attention has been given to biomechanical factors that may be associated with the development of overuse running injuries. Specifically, excessive pronation at the subtalar joint (or eversion) has been believed to be associated with various lower extremity injuries, as well as excessive tibial rotation (Areblad, Nigg, Ekstrand, Olsson & Ekstrom, 1990; Eng & Pierrynowski, 1994; McPoil & Cornwall, 1996; Hamill, van Emmerik, Heiderscheit & Li, 1999; Stergiou, Bates, & James, 1999; Bellchamber & van den Bogert, 2000; De Wit & De Clercq, 2000; Stacoff, Nigg, Reinschmidt, van den Bogert, Lundberg, Stussi, & Denoth, 2000; Stergiou, Houser, & Bates, 2000). Furthermore, internal tibial rotation is linked to knee flexion as external tibial rotation is linked to knee extension. It is generally accepted that the motion of the foot influences knee joint motion. It has been suggested that excessive tibial rotation may be the result of excessive foot pronation transferred, via the tibia, through a coupling mechanism at the knee (Stacoff, et al., 2000). Excessive internal rotation during the stance phase of running may delay the natural external rotation as the knee begins to extend (Bellchamber & van den Bogert, 2000). The delay in external rotation may be a mechanism by which torsional joint stresses at the knee or within the tibia may increase and, in turn, lead to knee injury. Thus, movement coupling between the ankle and the knee joint may be related with various knee injuries in running.

The mechanism of coordinative action between pronation and knee motion has received little attention in biomechanical research. Separately, rearfoot motion has been investigated extensively whereas tibial rotation has received very little attention, most likely due to the inability to measure this phenomenon without the use of three-dimensional (3D) kinematic analysis. In contrast, the majority of rearfoot motion studies have tended to use a two-dimensional (2D) approach. Recent literature has shown that a 2D analysis from a posterior view is significantly sensitive to the alignment angle between the foot and the camera axis and is also sensitive to the amount of foot abduction in individuals (Areblad, et al., 1990; McClay & Manal, 1998; McClay, 2000). Another suggested limitation to 2D analysis lies in the secondary planes of motion (McClay & Manal, 1999). McClay and Manal (1999) proposed that many of the abnormalities in running mechanics thought to be associated with injury occur in the secondary planes of motion (frontal and transverse). It was found that, although relatively smaller than the sagittal plane component, there was a substantial amount of positive power (generation) in the frontal plane at both the rearfoot and knee joints, which is ignored from a 2D perspective. Therefore, little is known about joint angular kinetics in the frontal and transverse planes regarding knee and rearfoot motion and its contribution to injury. Stergiou, et al. (1999) suggested that a possible mechanism responsible for various running injuries could be lack of coordination between the actions of the knee and the subtalar joint. Furthermore, they emphasized that increases in impact forces may augment this lack of coordination. Increased impact forces have been implicated as a cause of running injuries (Novacheck, 1998b; Stergiou, et al., 1999; McClay, 2000;

Morley, 2000; Scholten, 2000; Stergiou, et al., 2000). However, little is known about how increases in impact forces can cause running injuries. Stergiou, et al. (1999) used obstacle heights to produce increases in ground reaction impact forces (GRIF). The increases in impact forces resulted in a change in the rearfoot angle curve from a unimodal (one minimum) to a bimodal (two minimums) parabolic configuration. The appearance of a second minimum was attributed to a lateral deviation of the tibia, due to the increased impact with the ground. Although new evidence was presented regarding knee and rearfoot motion, the method of this study was two-dimensional; thus, it was sensitive to alignment error. Stergiou, et al. (2000) viewed the subtalar and knee joint relationship during running at various stride lengths. It was suggested that the longer the stride length, the greater the GRIF forces and the greater the rearfoot and knee angular velocity differences. A change in the rearfoot angle curve was also observed and was consistent with that of Stergiou, et al. (1999). However, the method of the study was also 2D and thus, sensitive to alignment errors. Stacoff, et al. (2000) investigated movement coupling between the calcaneus and tibia during the stance phase of running via a 3D analysis. It was found that movement coupling changed throughout the stance phase of running. The coupling coefficient between the calcaneus and the tibia was higher from heel-strike to mid-stance, where there is pronation and tibial internal rotation, when compared with mid-stance to toe-off, where there is supination and tibial external rotation. However, the method for kinematic analysis was based on intracortical bone pins. The external validity of the use of bone pins is questionable and reproducibility is difficult. Furthermore, it was noted that values of this study corresponded well with those of studies utilizing skin and shoe mounted markers, which gives support for the use of externally mounted markers. De Wit and De Clercq (2000) investigated the timing differences between subtalar and knee joint movement while running at variable speeds with shoes and while running barefoot. It was concluded that a large time discrepancy between knee extension and the end of pronation mainly depends on the presence of bimodal pronation curves. However, the method of the study was, again, 2D and there was a lack of kinetic data, which would give insight to differences in GRIF with increased speeds. Thus, with relatively little research concerning 3D analysis of the coordinative actions of the subtalar and knee joints, our understanding of their interactions is limited.

Purpose

The purpose of this study was to investigate the coupling mechanism between the subtalar and knee joints during running over obstacles of varying heights using a three-dimensional analysis.

Hypotheses

The following hypotheses were formulated:

1. As the obstacle height increased, it would cause an increase in ground reaction impact forces.
2. The increases in ground reaction impact forces would produce increased timing differences in the actions of the subtalar pronation/supination, knee joint flexion/extension, and tibial internal/external rotation. Differences in timing would be reflected in kinematic parameters.

3. The biomechanical evaluations and the clinical evaluations of subject rankings regarding susceptibility to injury would reveal a significant relationship.

Delimitations

The study was conducted within the following boundaries:

1. Ten male and/or female subjects between the ages of 18 and 35, free of injury and physical impairment at the time of evaluation, participated in the study. All subjects were required to be regular participants in physical activity at least 3 days per week.
2. Data analysis was limited to selected kinematic and kinetic parameters.
3. Biomechanical data collection consisted of one session per subject. A session consisted of running at a self-selected pace over obstacles of four different heights, including no height, for a total of five different conditions.
4. Each subject performed ten running trials per condition during the data collection session (50 total trials).
5. The orthopedic evaluation of the subjects was conducted on a separate day from the biomechanical data collection sessions.

Limitations

The following limitations were applied to the study:

1. Synchronization of the force (1000 Hz) and the video (240 Hz) data were constrained by the sampling rate of the video system. The maximum temporal discrepancy between the systems synchronized to the nearest video frame was estimated to be four milliseconds (ms) with an average error of two ms.

2. Limitations with the use of high-speed videography were acknowledged. These included digitizing errors (mean square error less than or equal to 5 millimeters) resulting from centroid location algorithms and movement of the identification markers placed on the skin of each subject.
3. Identification of joint centers was constrained by the application of externally mounted markers on the skin and shoes. These markers were estimations of the joint centers and anatomical landmarks, which were subjected to error from marker movement due to skin, adipose tissue, and muscle movement. Error may have also existed within shoe markers as the shoe was assumed to represent anatomical movement of the foot.
4. It was acknowledged that this study would not provide definitive answers as to why runners are injured.

Definition of Terms

Anterior-posterior force (F_y) – The component of the resultant ground reaction force (GRF) measured by a force platform in the anterior-posterior direction that occurs in the transverse plane.

Bimodal curve – A parabolic curve with two minimum or maximum values.

Center of pressure – The point of application of the resultant GRF vector on the force platform.

Coupling – the association of two systems in such a way that power may be transferred from one to another.

External tibial rotation – The rotation of the tibia in the transverse plane towards the lateral aspect of the body.

Foot-strike – The pattern of initial contact with the running surface by either the heel or the forefoot. It is usually associated with a toe-heel-toe footfall pattern during the stance phase of running.

Force platform – An aluminum plate mounted permanently in the floor that measures force output.

Ground reaction force (GRF) – The resultant force exerted by the ground to a subject in response to the force exerted by the subject to the ground; measured by a force platform.

Ground reaction impact force (GRIF) – The initial peak value of the vertical GRF curve resulting from the initial contact of the foot with the force platform.

Heel-strike landing – The pattern of initial contact with the running surface by the heel.

Internal tibial rotation – The rotation of the tibia in the transverse plan towards the midline of the body.

Joint angle – The angle formed between two adjacent segments.

Joint moment – The resultant moment effect produced by all forces (muscular, inertial, etc.) acting at a specific joint. Derived as the perpendicular distance from the estimated joint center of rotation and the ground reaction impact force.

Joint power – The product of joint moment and joint angular velocity. Determines the amount of eccentric and concentric work done at these joints.

Kinematic parameters – Descriptors of the spatial movement of the body and body segments derived from video record.

Kinetic parameters – Descriptors of force values as measured by the force platform.

Medial-lateral force (F_x) – The component of the resultant GRF measured by a force platform in a medial-lateral direction that occurs in the transverse plane.

Pronation – The triplanar rotation of the foot that places the foot into dorsiflexion, eversion, and abduction.

Segmental angular velocity – The rate at which a segment angle changes.

Support phase – The period of the running cycle where the body is supported by having one leg in contact with the ground; also referred to as the stance phase.

Toe-off – The instant at which the foot leaves the ground.

Touch down – The instant of the initial contact with the running surface by any structure of the foot.

Unimodal – A parabolic curve with only one minimum or maximum value.

Vertical force (F_z) – The component of the resultant GRF measured by a force platform in a direction perpendicular to the horizontal plane.

Significance of the Study

The primary objective of this study was to enhance our understanding of lower extremity function during human locomotion in a running state. With superior understanding, it becomes possible to uncover underlying mechanisms that may be implications as possible precursors to injury. The sagacity of these mechanisms may then provide the median for prevention of overuse injuries that occur as a result of running. Therefore, this study has the potential to benefit other disciplines, such as, neuroscience, orthopaedics, health practitioners, coaches, and any other discipline that has an interest in preventing foot injuries. Shoe construction, robotics, design of artificial limbs and general research on bipedal locomotion may also benefit from this study.

CHAPTER II: REVIEW OF LITERATURE

The following review of literature is divided into several sections and subsections, including 1) biomechanics of running, 2) running injuries, subdivided into impetus to run and specific running injuries, 3) lower extremity coupling, and 4) experimental design, subdivided into obstacles as a perturbation, 3D design, and clinical external validity. A summary completes the chapter.

Biomechanics of Running

To understand the mechanisms of running injuries, a basic understanding of the biomechanics of running is essential. Running is a bipedal, alternating rhythmic form locomotion. Running contains all the determinants of walking, but to a greater degree, and includes such entities as pelvic rotation, pelvic tilt, lateral movement of the pelvis, flexion and extension of the legs and arms, and pronation and supination of the ankle. The demarcation between walking and running occurs when periods of double support during the stance phase of walking shifts to two periods of double float at the beginning and the end of the swing phase of the gait cycle.

The gait cycle (Figure 1) is the basic unit of measurement for analysis of human locomotion. The gait cycle begins when one foot comes in contact with the ground and ends when the same foot contacts the ground again, or vice versa (Thordarson, 1997; Novacheck, 1998a). These two instances of foot contact are referred to as initial contact or foot strike. The gait cycle is divided into two phases, the swing phase and the stance phase. The stance phase (Figure 2) occurs when the foot is in contact with the ground and, conversely, the swing phase occurs when the foot is not in contact with the ground

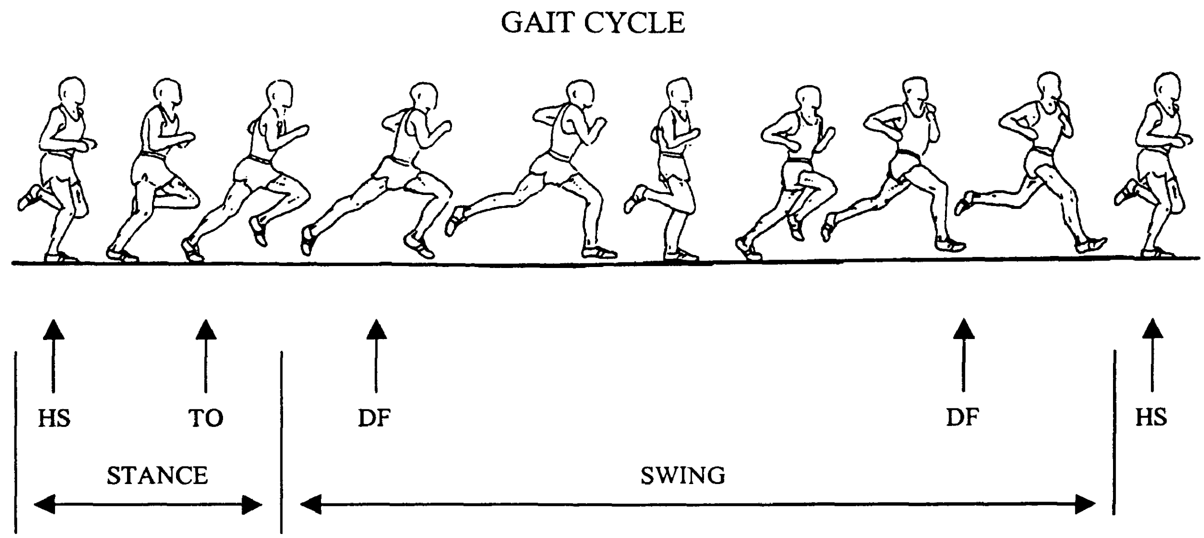

Figure 1. The gait cycle in running. HS = heel-strike, TO = toe-off, DF = double float. (Slocum & James, 1958).

(Thordarson, 1997). The beginning of stance is marked by initial contact. The initial contact during running is typically a heel strike pattern where the heel of the foot is the first to contact the running surface (Adrian & Cooper, 1995: Scholten, 1999). The foot then progresses to full contact with the running surface, as the toe is the last element of the foot to contact the running surface. Scholten (1999) and McClay (2000) mentioned that the majority (75 to 80%) of runners uses this heel-strike pattern. The end of stance is characterized by toe-off, where the toe is the last element to leave the ground. Toe-off also signifies the beginning of the swing phase and it occurs at approximately 39 to 40% of the gait

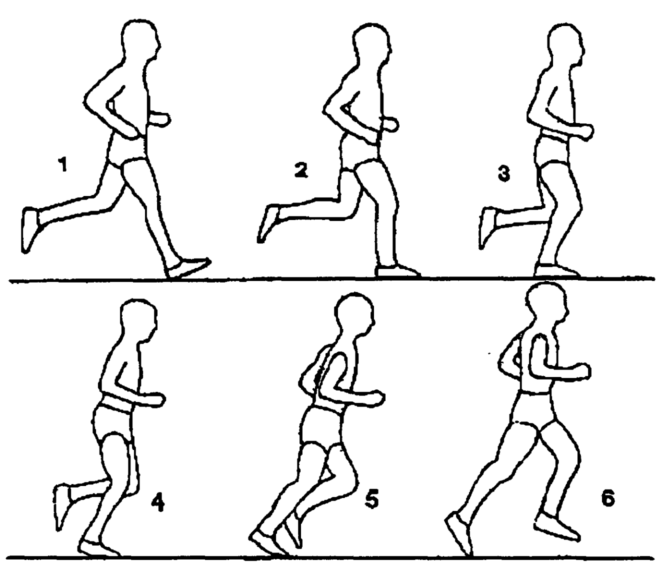

Figure 2. Selected positions of the stance phase of the gait cycle. 1 = heel-strike, 6 = toe-off (Subotnick, 1999, 173).

cycle (Thordarson, 1997; Novacheck, 1998b). The remaining 60% of the gait cycle is spent in the swing phase. During running, there are no periods when both feet are in contact with the ground at the same time. Instead, during the swing phase, both feet are in the air twice; a brief instance at the beginning of swing, and a similar brief instance at the end of swing (Thordarson, 1997; Novacheck, 1998b). The two airborne periods are referred to as double-float.

The stance phase (Figure 3) can be subdivided into two periods of alternating deceleration and acceleration of the body, referred to as absorption and generation respectively. During the period of absorption, the body's center of mass falls from its peak height during double-float to its minimum height, which occurs at maximum knee flexion. During the period of generation, the body's mass is propelled upward and forward. As can be seen, these periods do not coincide with the incidence of heel-strike and toe-off. It is generally thought that most running injuries occur during stance and the absorption period of the gait cycle (McClay, 2000).

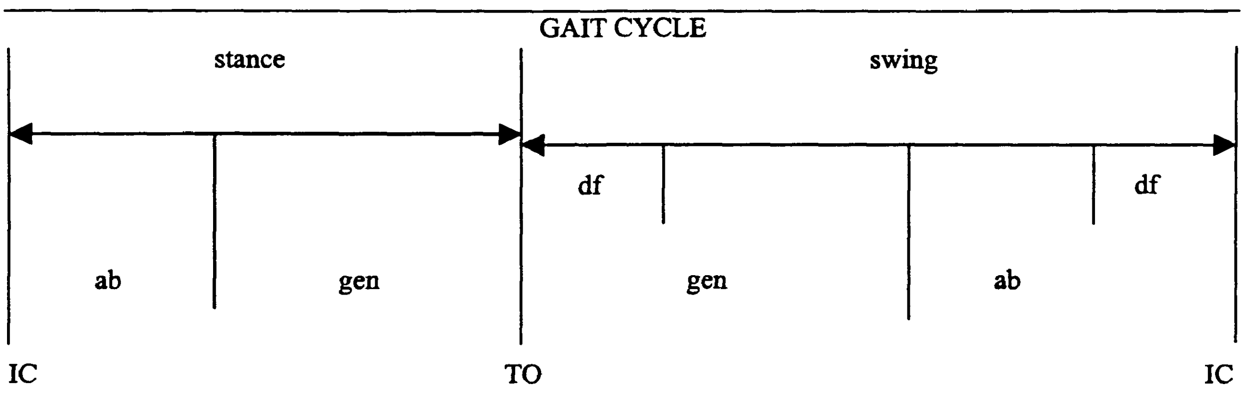

Figure 3. The events of the running gait during one gait cycle. IC = initial contact, TO = toe-off, ab = absorption, gen = generation, and df = double-float.

Running Injuries

Impetus to Run

The physical activity of running has matured to a state that is considered to be an important component to overall fitness and longevity. Running as a physical activity is capable of addressing many of the goals and objectives in the Healthy People 2010 report issued by the Office of Disease Prevention and Health Promotion (U.S. Department of Health and Human Services, 2000). Due to the weight-bearing nature of running, it is recognized by the Surgeon General's Report on Physical Activity and Health (1996) as having the capability to increase bone strength and delay the onset of frailty as people age and protect against disorders, such as osteoporosis. However, running injury free and benefiting from these positive attributes can be very difficult.

Novacheck (1998b) has reported that approximately 30 million Americans run for recreation or competition. Fields (1994) estimated that running has 50 million devotees who choose this activity for their primary method of fitness. Walker (1998) simply states that, "Millions of Americans are runners." These numbers have increased significantly since the 1970's when running as a recreational activity first began (McClay, 2000). With the increasing number of running participants, there has also been an increase in the number or runners who sustain an injury each year. Novacheck (1998a) estimated that each year 25 – 50% of runners will sustain an injury that is severe enough to cause a change in practice or performance. Walker (1998) estimates that over half of all runners will sustain an injury each year. Brill and Macera (1995) quote a yearly incidence rate of approximately 30 – 65%. van Mechelen (1992) and Wen et al. (1996) quoted similar approximations of 37 – 56%. Finally, Browning and Donley (2000) report that at least 25% of adults sustain a running injury each year.

Specific Running Injuries

Approximately 70-80% of all running injuries occur in the lower extremities (Brill & Macera, 1995; Novacheck, 1998b; Stergiou, Bates, & Houser, 1999). The primary genesis of all running injuries are overuse or overload type injuries, affecting 50-80% of all runners (Fields, 1994; Brill & Macera, 1995; Fredericson, 1996; Hintermann & Nigg, 1998; Walker, 1998; Stergiou, et al., 1999; McClay, 2000). Over half (50-60%) of all running injuries affect the musculoskeletal system (Fields, 1994; Browning & Donley, 2000). The most common site for lower extremity, overuse, musculoskeletal injuries is the knee, followed by the lower leg and the ankle (McClay, 2000).

The knee is the most prevalent site of running injuries. Hamill, et al. (1999) quoted work by MacIntyre, Taunton, Clement, Lloyd-Smith, McKenzie, and Morrell (1991) where it was reported that over 25% of all running injuries occur at the knee. Fredericson (1996), Novacheck (1998a), and Browning and Donley (2000) estimated that 30 to 50% of all running injuries occur at the knee in the form of anterior knee pain. Anterior knee pain includes patellofemoral pain (stress) syndrome and chondromalacia; both are often referred to as "runner's knee" (Browning and Donley, 2000). Patellofemoral pain syndrome is a general term for pain at the patellofemoral articulation, where irritation around or behind the patella may occur (Fields, 1994). Chondromalacia is a specific condition in which the articular cartilage is softened and fibrillated (Walker, 1998; Browning and Donley, 2000). Predisposing factors for abnormal tracking of the patellofemoral joint include torsional or angular malalignment of the lower extremity, including external torsion of the tibia and subtalar pronation of the foot (Fredericson, 1996). Another common knee injury, other than anterior knee pain, is iliotibial (IT) band syndrome. This syndrome produces lateral knee pain that results from friction of the IT band moving over the lateral femoral condyle with repetitive knee flexion beyond 30 degrees (Fields, 1994; Fredericson, 1996).

Injuries to the lower leg include tibial stress fractures and medial tibial stress syndrome (shin splints). Runners commonly experience pain along the medial border of the tibia, which may be due to tibial stress fractures or shin splints. Walker (1998) estimates that tibial stress fractures account for 15 – 18% of running injuries. These injuries are the consequence of the failure of bone to successfully adapt to the encountered forces applied repetitively during running. Runners strike the ground approximately 600 times per kilometer and with every heel-strike, a peak vertical GRF of two to four times the runner's body weight is applied to the leg (Crossley, Bennell, Wrigley, & Oakes, 1999). The primary distinguishable symptom between tibial stress fractures and shin splints is that tibial stress fractures are associated with localized pain that persists after running and it continues with daily ambulation (Fredericson, 1996). Also, pain associated with shin splints generally occurs along the medial, distal two-thirds of the tibia (Walker, 1998). For unclear reasons, women tend to have a slightly higher incidence of stress fractures than men do (Fredericson, 1996; Walker, 1998).

Injuries to the foot and ankle include Achilles tendonitis and plantar fasciitis. Walker (1998) estimates that Achilles tendonitis and plantar fasciitis each comprise 10%

of all running injuries. Achilles tendonitis is an inflammation of the Achilles' tendon within the mid-substance of the tendon and near its insertion to the calcaneal tuberosity (Walker, 1998; Browning & Donley, 2000). The tendon is subjected to large stresses that can vary between 2000 and 7000N (Fredericson, 1996). This is the equivalent to 10 times the runner's body weight and is repeatedly exerted with each heel-strike. Furthermore, since the Achilles inserts on the calcaneus, motion at the subtalar joint will place a rotational force on the tendon fibers. For example, the pronated foot normally imparts an internal rotation force on the tibia, whereas knee extension imparts and external rotation force through the tibia. Fredericson (1996) and Browning and Donley (2000) proposed that if the foot remains pronated excessively as normal knee extension occurs, the Achilles experiences abnormally high forces resulting from these contradictory rotational forces. Plantar fasciitis is the most troublesome and the most common cause of inferior heel pain in runners (Fields, 1994; Fredericson, 1996; Browning & Donley, 2000). This condition is the result of inflammation due to chronic stretching, irritation, and microtearing of the plantar fascia. The plantar fascia is a band of connective tissue extending from the calcaneal tuberosity to the metatarsal heads of the foot. Inflammation is thought to occur due to the repetitive stress to the fascia at its origin (Fredericson, 1996; Browning & Donley, 2000).

In summary, approximately half of all runners will sustain a running related injury each year. The majority of these running injuries are of the overuse kind, affecting the musculoskeletal system of the lower extremities. The primary overuse injury occurs at the knee and includes anterior knee pain and IT band syndrome. Overuse injuries to the lower leg and foot and ankle are common as well and include shin splints, tibial stress fractures, Achilles tendonitis, and plantar fasciitis. The vast majority of lower extremity running injuries appears to be related to cumulative overload resulting from the repetitive, rhythmic impact with the running surface. During running, the repetitive and violent encounter between the foot and the running surface results in impact forces that have to be absorbed by the supportive structures. The joint actions of knee flexion and subtalar pronation are both associated with attenuation of these impact forces (Stergiou, et al., 1999).

Lower Extremity Coupling

The focus of running studies to date has been on the individual actions of lower extremity joints. However, the primary lower extremity joints (ankle, knee, hip) are all linked mechanically, thus, motions at the foot will influence movements at the tibia, and the tibia will then influence movements at the femur. Recently, some attention in the literature has been given to the interactions among these joints.

As stated previously, the majority of runners strike the ground in a heel-to-toe pattern and with each strike or impact, two to four times the runner's body weight may be transmitted through the lower extremity. This shock must be attenuated throughout the body, and absorption of this shock does not occur instantaneously. Instead, it is dissipated during the first half of stance through several different tissues and supportive structures (Novacheck, 1998b; Stergiou, et al., 1999). The joint actions of knee flexion and subtalar pronation are both associated with attenuation of these impulsive shock loads (Hintermann & Nigg, 1998; Stergiou, et al., 1999; Stergiou, et al., 2000). Kim, Voloshin, and Johnson (1994) quoted that 70 to 80% of impact peak reduction could be credited to the knee joint. At the distal end of the tibia, subtalar pronation allows for the impact forces to be absorbed within the supporting structures by increasing the time with which the foot is in contact with the running surface (Stergiou, et al., 1999; Stergiou, et al., 2000). Without subtalar pronation, these forces would have to be abruptly and directly absorbed by the supporting structures.

Bates, James, and Osternig (1978) were the first to suggest that a possible relationship may exist in the coordination of these joints and injury. Recent attention has been given to this thought that a disruption in the coordination of movement between the subtalar joint and the knee joint may result in injury (Hintermann & Nigg, 1998; Novacheck, 1998b; Stergiou, et al., 1999; Bellchamber & van den Bogert, 2000; De Wit & De Clercq, 2000; McClay, 2000; Stergiou, et al., 2000). The subtalar joint has been modeled as a mitered hinge joint where rotation of the foot about the longitudinal axis is transmitted to the tibia, imposing rotation of the tibia through its longitudinal axis (Stergiou, et al., 1999). During running, pronation begins at heel-strike, causing rotation of the foot as the calcaneus everts and the foot dorsiflexes and abducts, and ends at approximately midstance (Hintermann & Nigg, 1998). The calcaneus eversion induces internal tibial rotation via the subtalar joint. In the tibiofemoral joint, the medial condyle extends further than the lateral condyle by 1.7 cm distally, which causes the tibia to rotate around the femur during flexion and extension (Nordin & Frankel, 1989). Specifically, as the knee flexes, the tibia rotates internally around the femur and as the knee extends, the tibia rotates externally. Thus, tibial rotation is coupled to pronation of the foot through the subtalar joint and is also coupled to the knee through the tibiofemoral joint.

The subtalar joint axis is reported to be inclined in the sagittal plane approximately 45 degrees from the bottom surface of the foot (Morris, 1994; Bellchamber & van den Bogert, 2000; McClay, 2000). In theory, as the calcaneus everts, there should be an equal amount of tibial internal rotation. Bellchamber and van den Bogert (2000) quoted work done be Nigg, Cole, and Nachbauer (1993) where the correlation between the coupling of foot eversion and internal tibial rotation was r^2 = 0.991 for running. The transition between knee joint flexion to extension and subtalar joint pronation to supination generally occurs at approximately the same time during midstance (Stergiou, et al., 1999; Bellchamber & van den Bogert, 2000; De Wit & De Clercq, 2000; McClay, 2000). However, if the subtalar joint begins the transition at a different time than the knee joint, an antagonistic relationship may develop. In other words, if more time is spent in pronation and normal knee extension occurs, there will be an internal rotation at the distal end of the tibia coupled with external rotation at the proximal end of the tibia. Likewise, if less pronation occurs and supination begins before knee extension begins, there will be external rotation occurring at the distal end and internal rotation at the proximal end. If this antagonistic relationship occurs repeatedly, as is the case with every foot strike during running, increased stress on the soft tissues of the knee or increased torsional loads within the tibia itself could result in injury (Stergiou, et al., 1999; McClay, 2000). Likewise, abnormal torsional loads are considered to be a possible cause of osteoarthritis at the knee (Stergiou, et al., 1999;

McClay, 2000). It should also be noted that many of the most common overuse injuries in running are believed to be associated with the transfer of movement in the frontal plane (eversion) to transverse plane movement (tibial rotation) (McClay, 2000).

In summary, few studies exist that investigated the coupling of the lower extremity joints. During the stance phase of running, there is a coordination pattern that exists between pronation and knee flexion in that the transition from pronation to supination should occur at the same time as the transition from knee flexion to knee extension. Asynchrony may develop in this coordination by way of increased or decreased time spent in pronation. With asynchrony in the coordination pattern occurring during every stance phase experienced during running, there exists a possible mechanism for abnormal stresses to develop and to induce injuries at the knee and foot.

Experimental Design

Obstacle Height as a Perturbation

The effects of obstacle height on kinematics and kinetics have not been examined as well as other perturbations, such as speed (Stergiou, 1995; Scholten, 1999; Stergiou, et al., 1999). This is somewhat surprising considering the rough terrain that many runners encounter on a consistent basis. Several studies exist where subjects are instructed to walk over obstacles of varying height (Chou & Draganich, 1997; Chou, Daufman, Brey & An, 1998; Houser, 1999). However, the primary focus of these studies concerned the contralateral limb (swinging leg) and the perception relative to the obstacle's height and the different strategies used to clear it. Two studies exist where subjects were instructed to run over obstacles of varying height. Stergiou, et al. (1999) required subjects to run at a previously self-selected pace over obstacles of three different heights: 5%, 10%, and 15% of their standing height. The obstacles were placed directly before the force platform so that the subject had to clear the obstacle with the right leg and land on the force platform. Stride length was also controlled for by placing a marker one step before the force platform to identify left foot position. The subjects were instructed to hit the marker with their left foot before clearing the obstacle with the right leg. A similar procedure was utilized by Scholten (1999) in that a similar obstacle was cleared while running at heights of 10%, 12.5%, 15%, 17.5%, 20%, and 22.5% of a subject's height. Stride length was controlled in a similar fashion as well. However, in the study by Scholten (1999), subjects were instructed to run over the obstacle and land in a manner that felt comfortable, whereas Stergiou, et al. (1999) instructed subjects not to "jump" over the obstacle in order to ensure the typical heel-to-toe landing. Both studies revealed that as obstacle height increased, ground reaction forces increased as well. Scholten (1999) also found that between the 12.5% to 15% obstacle conditions, the group response transitioned from a heel-to-toe foot strike pattern to a toe-to-heel pattern. It was speculated that this new pattern was utilized to help absorb the increased impact forces, which would have increased the involvement of the ankle joint in shock absorption.

Three-dimensional Design

Studies analyzing lower leg motion and rearfoot motion during running have been traditionally using a 2D analysis (Knutzen & Price, 1994; Scholten, 1999; Stergiou, et al., 1999; De Wit & De Clercq, 2000; Morley, 2000; Stergiou, et al., 2000). Recently, however, there have been more studies where a 3D analysis is being utilized (Areblad, et al., 1990; Eng & Pierrynowski, 1994; Eng & Winter, 1994; McClay & Manal, 1998; McClay and Manal, 1999; McClay, 2000). Areblad, et al. (1990) proposed that description of 3D motion between the foot and the lower leg could be made in at least three principally different ways:

> Firstly, the rotations can be measured about the ankle and subtalar joint if the positions of these axes are known. Secondly, the rotations can be measured about the clinical axes describing eversion/inversion, dorsi-/plantar flexion and ab-/adduction. Thirdly, the motion can be described as rotation and translation about a screw axis, which does not have any anatomical correspondence.

The results of a 2D technique are affected by projection errors, which depend on the alignment of the segments with the plane being filmed (Areblad, et al., 1990). Areblad, et al. (1990) found that the initial rearfoot angle and the change in pronation of the rearfoot angle strongly depended upon the alignment angle, predicting a lower angular value with a more abducted foot. A change of 2 degrees in the alignment angle resulted in an approximate change of 1 degree in the computed angle. McClay and Manal (1998) found that when the frontal plane of the foot moves out of the frontal plane of the lab, as with foot abduction, the 2D eversion angle underestimates the true 3D eversion angle. Significant differences ($p<0.01$) were found for 2D and 3D eversion at heel-strike (2D = -1.9^0 eversion, 3D = -3.5^0 eversion), eversion at toe-off (2D = 10.4^0 inversion, 3D = -1.6^0 eversion), and time to peak eversion (2D = 0.106 s, 3D = 0.092 s). The 2D/3D differences were greater for subjects with excessive foot abduction, although these differences were small (less than 2 degrees) until foot abduction exceeded 20 degrees. This is most likely to occur on subjects that have excessive foot abduction or while plantarflexing at late stance. It was concluded that caution should be exercised in interpreting 2D rearfoot variables at foot strike and toe-off, and those involving times to peak values, especially for subjects with excessive foot abduction (McClay & Manal, 1998).

Aside from the susceptibility to perspective errors, 2D motion analysis techniques give no insight to joint kinetics for the frontal and the transverse planes of movement. The study of angular joint kinetics is of importance as it can provide an understanding into the function of muscles involved. It can also provide information pertaining to the joint loads experienced by the lower extremity. When loads become excessive, the risk for injury may increase. An estimation of the joint moments may be obtained through inverse dynamics. Inverse dynamics is a series of mathematical equations that incorporates joint positions (acquired via motion analysis), ground reaction force data, and anthropometric data to calculate joint moments (Novacheck, 1998b). Although it is muscle forces that cause the body to accelerate, these forces are difficult to measure directly. However, the acceleration of the body can be measured and force and moments are then derived mathematically from the ground up; hence, the process is termed inverse dynamics (McClay, 2000). Moments alone do not explain the manner in which muscles are functioning. Joint power is then determined by multiplying joint moments by the corresponding angular velocities. This information gives insight to the amount of eccentric and concentric work done during running. However, the clinical significance of these variables is not well understood (McClay, 2000). McClay and Manal (1999) found that, although relatively smaller than the sagittal plane component, a substantial amount of positive power or generation was done in the frontal plane at the knee joint (18.9%)

and at the subtalar joint (16.1%) during running. Transverse plane kinetics were extremely variable at these joints. For the ankle, an abduction moment (outward rotation) was initially seen followed by an adduction moment (inward rotation) and ending with an abduction moment. This pattern was the most variable of all ankle variables. Only 6.4% of the total negative power (absorption) and 0.7% of the total positive power (generation) was seen in the transverse plane. Transverse plane knee kinetics showed similar results to the ankle (McClay & Manal, 1999). It should be noted that many of the injuries that are believed to be related to abnormal mechanics are most likely associated with deviations in the frontal and transverse planes.

In summary, utilizing 2D motion analysis is subjected to perspective errors and does not give complete insight to joint kinetics. It was found that 2D variables measured from a posterior view are very sensitive to the alignment angle between the foot and the camera axis. Also, caution should be exercised when assessing a 2D rearfoot motion in subjects with excessive foot abduction.

Clinical External Validity

The medical community suggests that there has been little progress made in our understanding of the mechanisms of running injuries, which hinders the ability to prevent injury (Fredericson, 1996; McPoil & Cornwall, 1996; Novacheck, 1998a; Stergiou, et al., 1999). A major problem concerning the prevention of running injuries is the inability of clinical evaluations to validate biomechanical results. In an attempt to increase the clinical external validity of their results, Stergiou, et al. (1999) constructed subject rankings for susceptibility to injury based on a clinical evaluation and the biomechanical data. In that study, the biomechanical subjects' rankings were based on the variable ability of every subject to sustain proper coordination through the changes introduced in the experiment, i.e. variable obstacle heights and speed changes. Proper coordination was defined as the ability of a subject to maintain time matching and sequencing between the actions of subtalar pronation/supination and knee flexion/extension. Lack of coordination was assumed to be an injurious situation. The subject's scores from two rankings, curve correlations and velocity differences, were added and the sums were used to construct a final ranking. The subject with the lowest score was assigned a rank of one and vice versa. The clinical evaluation was constructed based on a clinical evaluation by a well-established orthopedist and sports medicine specialist. The specialist ranked the subjects on susceptibility to injury based on injury history, running experience, and clinical examination. It was noted that this method was not based on previous research and it was entirely subjective. However, the evaluation of patients in the clinical setting is currently based on the same method, the specialist's perception of symptoms' importance. It was revealed that the velocity differences, which provide a reasonable estimate of the antagonistic relationships between the actions of the subtalar and the knee, produced the highest correlation with the clinical evaluation ($r = 0.798$). Therefore, the results of their study, and especially the parameter of velocity differences as a biomechanical tool, was a serious attempt to bridge the gap of information exchange and assist in the evaluation of running injuries (Stergiou, et al., 1999).

Summary

Substantial research attention has been given toward enhancing the understanding of the mechanisms by which runners are injured. Previous research has found little correlation between specific anatomic abnormalities and abnormal biomechanics of the lower extremity with specific running injuries. This lack of correlation could be the result of the inability to incorporate all the various functional components into study designs. In addition, with the advent of three-dimensional motion analysis, the ability to decrease measurement error and properly investigate important components is tremendously enhanced.

CHAPTER III: METHODS

Subjects

The subjects of this investigation consisted of a total of 10 healthy male and female runners between the ages of 19 and 35. Subject demographics can be found in Appendix C. All subjects were recreational runners, participating in running for a minimum of three days per week, for at least one year. Each subject was free of injury and physical impairment at the time of testing. All subjects were heel-strikers, which was determined before testing. Also before testing, each subject was required to read and sign an informed consent form approved by the Institutional Review Board of the University of Nebraska at Omaha.

Instrumentation

All biomechanical data collection took place at the University of Nebraska at Omaha Biomechanics Laboratory. The biomechanics lab is located in the Health Physical Education and Recreation building and measures 110 feet by 70 feet with ceilings measuring 20 feet in height. This provided ample room for subjects to perform the running activity. The floor is a multipurpose activity floor, which provides a safe running surface.

Ground Reaction Force Data

A Kistler multi-component force platform (Model 9281-B11) mounted flush with the multipurpose laboratory floor, midway along a 30-meter runway, was utilized to collect ground reaction forces. The force platform is mounted on a stainless steel base plate secured to a concrete cylinder that is an integral part of the building substructure.

The force platform is calibrated on a yearly basis using known weights. The Kistler force platform system utilizes piezoelectric transducers and measured forces at 960 Hz. The quartz crystals deform and generate electrical outputs or voltage. These voltages are then amplified with a Kistler signal conditioner/amplifier (Model 9865B). The forces are then displayed in the three cardinal directions: vertical (F_z), anterior/posterior (F_y), and medial/lateral (F_x). The signal conditioner/amplifier was connected to a 16 channel Peak Performance Technologies Analog/Digital Interface Unit (Peak ADIU: Model 2051), which converted the voltages to digital units for the computer to read. The ADIU was interfaced to a computer equipped with a Pentium II processing chip and containing 'Peak Performance Technologies' 'Peak Motus 4.3.1 System' software. The Peak system compiled the digital data for future processing.

Videography

Two JC Labs Inc. 240 Hz cameras (Model HSC250) were utilized to capture kinematic data. The cameras were mounted at a height of approximately 2 m and approximately 7 m apart. Camera one (master) was positioned 9.9 m from the center of the force platform while camera two (slave 1) was located 10.3 m from the center of the force platform (Figure 5). Each camera was equipped with a zoom lens to optimize image size in order to minimize perspective error.

Before the onset of data collection, on-site accuracy tests were completed. The space where 3D motion occurred was defined in terms of a Cartesian coordinated system. This system is comprised of three, perpendicular axes, X, Y, Z, with a common origin. The projections of a point on the X, Y, and Z axes are the real-space coordinates: x, y, and z. To locate an object in space, a minimum of two cameras is required. Each camera describes the real-space coordinates on a 2D plane and each camera has different coordinates for the same point. A calibration frame with several known real-space coordinates, referred to as control points, defines an arbitrary origin, the orientation of the axes, and the scaling of a real-space Cartesian coordinate system. A Peak Performance 25-point, 9-rod calibration star (Model D1), reduced to the inner 17 points, was used to calibrate space. This calibration frame was placed such that all balls (markers) on the rods were in view of all cameras and, thus, approximated the volume of space where the motion occurs. All points that were generated during data collection required known camera constants and coordinates to reconstruct real-space coordinates, and this was performed using Direct Linear Transformation (DLT). The DLT method establishes a direct linear relationship between the digitized 2D coordinates from the two cameras and the 3D space coordinates from the control frame. In addition, a Global Transformation frame was used to define three points, which translated and rotated the axis of the coordinate system. A level of accuracy was then determined based on the digitized control frame by computing the Root Mean Square (RMS) of the difference between known control point coordinates and the digitized coordinates. A RMS of less than five millimeters was used as an acceptable error accuracy, with which data collection may begin. Once accuracy is deemed acceptable, the cameras did not change their focal setting throughout data collection.

Reflective markers were placed on the right lower extremity of each subject to identify the following landmarks: 1) top of shoe above 2nd metatarsal head, 2) bisection of back of shoe at mid-calcaneus, 3) lateral malleolus, 4) tibial tuberosity, 5) lateral femoral condyle of the knee, 6) mid-thigh, and 7) greater trochanter of the hip. Placement of the reflective markers is based on the model developed by Vaughan, Davis, and O'Conner (1999). The model by Vaughan, et al. (1999) selects three markers to model a segment. Once a segment is created, an orthogonal reference system based on these three markers is created. Prediction equations, based on anthropometric measurements and the orthogonal reference system, then estimates the joint center positions. In the model that was used for this study, the foot was constructed by using the 2nd metatarsal, mid-calcaneus, and lateral malleolus markers, the shank was created by the lateral malleolus, lateral femoral condyle, and tibial tuberosity markers, and the thigh was created by the lateral femoral condyle, greater trochanter, and mid-thigh markers (Figure 4). In addition to the anatomical markers, a marker was placed on the obstacle to assist in determining the location of the obstacle in the field of view. The reflective markers were round, Styrofoam balls encased in 3M reflective tape, measuring 25 mm in diameter, and were illuminated by two Pallite VII

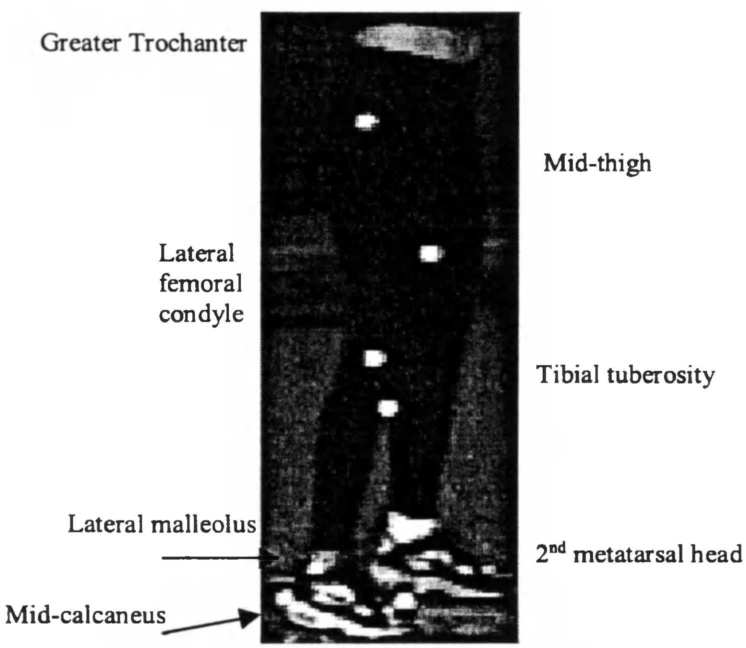

Figure 4. 3D marker set-up based on Vaughn et al. (1999). Each marker consists of a Styrofoam ball, 25 mm in diameter, encased in 3M reflective tape.

lamps, each containing four, 300 W bulbs for a total of 2400 W. Each Pallite was located directly under a camera for optimal reflection.

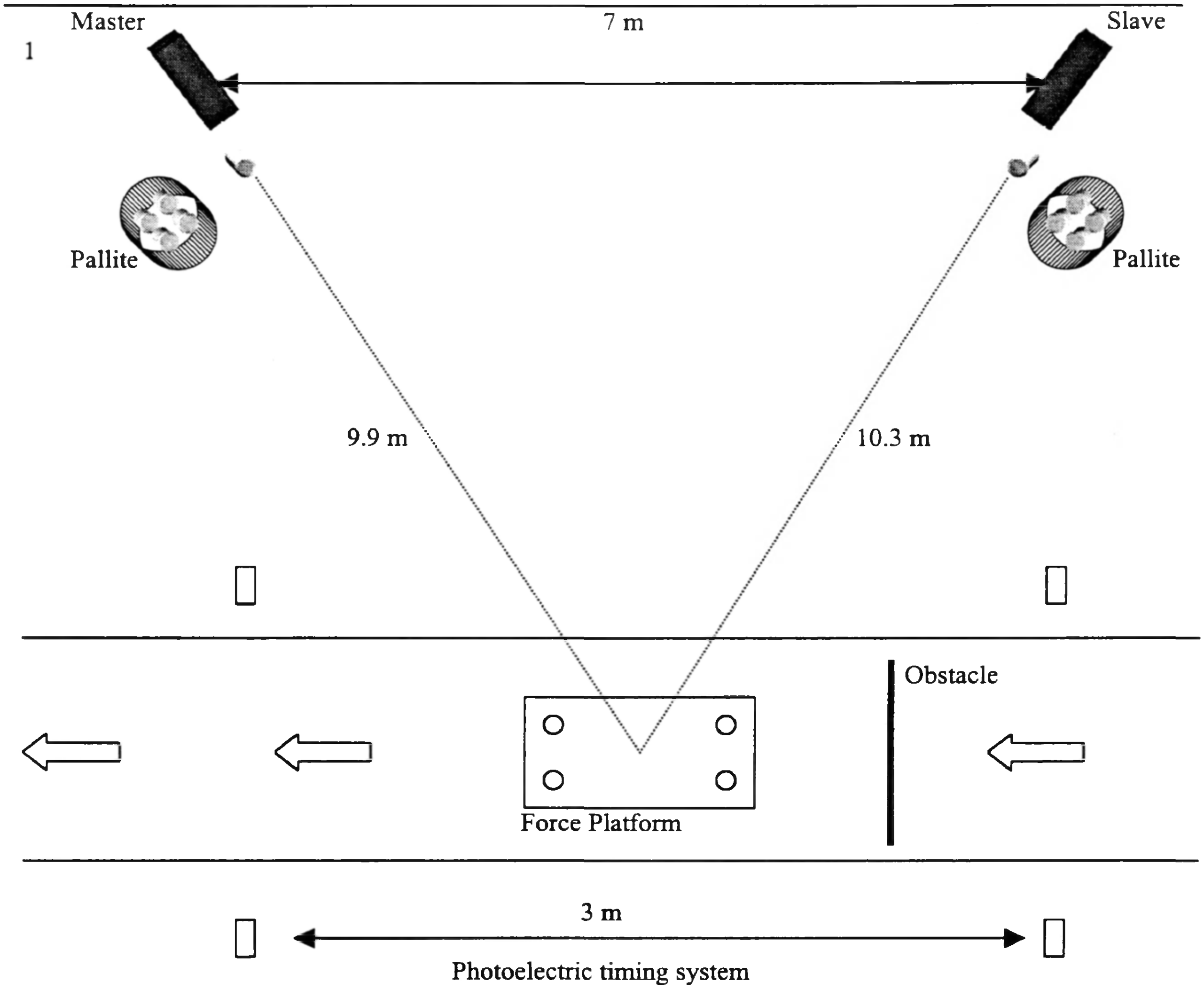

Figure 5. General layout of laboratory with the path of the subject indicated by the block arrows. The dashed lines indicate the field of view of each camera.

The video images obtained from the cameras were stored on two SVHS videotapes via two Panasonic SVHS VCRs (Model AG-1970P). The VCRs were interfaced with a Magnavox TV fitted with a video switch for instant qualitative evaluation of the video recording from both cameras. Collection and synchronization between the GRF and video data was accomplished via a manual thumb switch. The manual switch was connected to a Peak Event Synchronization Unit (ESU). The Peak ESU was interfaced with both VCRs and the Peak ADIU. When the manual switch is pressed, it generates a voltage pulse square wave that initiates GRF data collection and is sampled as a synchronization channel. Simultaneously, it places a vertical 16-line digital bare code in the upper left corner of the video picture. The initial frame when the bar code appears on the video picture and the time of initiation of the square wave was used to synchronize the video and force data.

The video data was transformed to digital format and digitized via the Peak Motus video system. The video images recorded on the SVHS videotapes were imported into the computer, where they were split into two fields and presented to the operator on the video monitor with a superimposed cursor. The cursor can be manipulated with an optical mouse to identify x, y coordinates. The coordinate data was extracted in ASCII format files for subsequent analysis.

Experimental Protocol

Subjects attended two test sessions on two different days. Testing took between one to one-and-a-half hours for the clinical exam and between two to three hours for the biomechanical data collection. On the first day, subjects underwent an orthopedic examination by a licensed physical therapist (Jason Paladino, MPT, ATC). The physical therapist administered a battery of orthopedic tests to evaluate lower extremity alignment and geometry. The tests were indicated on the Informed Consent Form (Appendix A) and can be found in Appendix B. On the second test day, subjects performed an obstacle manipulation protocol. Subjects wore their regular running shoes to assure the most normal performance. Minimal clothing was worn to improve the identification of anatomical landmarks for the positioning of the reflective markers.

Speed was monitored over a three-meter interval using a photoelectric timing system (LaFayette Performance Pack, Model-63520) (Figure 5). Subjects were given time to accommodate to the experimental set-up and to adequately warm-up prior to testing. Warm-up consisted of running through the testing area without concern for stepping on the force platform. During warm-up, the subject established a comfortable running pace, and a range of five percent above and below this pace was recorded by the investigator. Subjects then ran within their previously self-selected pace range over obstacles of four different heights: 5%, 7.5%, 10%, and 12.5% of the subjects standing height. A condition with no obstacle was utilized as well. The presentation of obstacle heights was randomized to eliminate order effect. The selection of the obstacle heights was established based upon the related running literature (Stergiou, 1995; Stergiou, et al., 1999; Scholten, 1999). The obstacle heights that were used in this study were typical of those encountered when traversing a normal varied terrain. The obstacle was placed before the force platform so that the subject would have to clear the obstacle with the right leg and land on the force platform. Placement of the obstacle was determined by identifying the subject's left foot contact and positioning the obstacle halfway between the middle of the force platform and left foot contact. A non-reflective marker was used to identify left foot position. Subjects were instructed to run naturally over the obstacle and not to jump over the obstacle to ensure a heel-strike-landing pattern. If a subject jumped over the obstacle, the trial was thrown-out and another trial was performed. When the obstacle was placed on the runway, the subjects were instructed to hit the marker with their left foot prior to clearing the obstacle with the right leg. This procedure ensured that the subject did not change their stride length when clearing the varying obstacle heights. The obstacle was made of extremely lightweight wood so that if a subject stepped on or hit the obstacle by mistake while running, the obstacle collapsed. This minimized the risk of the subject tripping and/or falling. Each obstacle condition consisted of 10 good trials. A good trial was considered a trial where the subject cleared the obstacle without jumping over the obstacle, landed on the force platform in a typical heel-strike pattern, and was within the self-selected running pace range. The total number of trials for the testing session was 50 good trials.

Data Analysis

Data Reduction: the first peak value (impact peak) of the vertical GRF (F_z) was identified for each trial. The GRIF values were then normalized for body mass and mean values were calculated across trials for each condition of each subject. The group means of all subjects were also calculated for each condition.

The kinematic coordinates were scaled and smoothed using a Butterworth low-pass filter with a selective cut-off algorithm. The cut-off frequency values were 13-16 Hz for sagittal plane coordinates and 16-20 Hz for the frontal and transverse plane coordinates. Stergiou, et al. (1999) determined that frequencies below 15 Hz for running would severely attenuate the impact phenomena (high frequency), thus resulting in an inadequate, over-smoothed curve. All kinematic parameter data files were normalized to

100 points for the support period using a cubic spline routine to enable mean ensemble curves to be derived for each condition of each subject.

Discrete Point Analysis: dependent variables from selected kinematic parameters included maximums, minimums, and ranges. Maximums, minimums, and ranges were determined for knee flexion/extension angular displacement in the sagittal plane, tibial internal/external rotation angular displacement in the transverse plane, and subtalar eversion/inversion angular displacement in the frontal plane. The respective times of occurrence were also identified for each trial. The respective times of occurrence are named as, time to maximum knee flexion (TMKF), time to maximum pronation (TMP), and time to maximum tibial internal rotation (TMTIR). In addition, the absolute differences between TMP and TMKF (|TMP-TMKF|), TMP and TMTIR (|TMP-TMTIR|), and TMKF and TMTIR (|TMKF-TMTIR|) were also identified for each trial. The mean values for TMP, TMKF, TMTIR, |TMP-TMKF|, |TMP-TMTIR|, and (|TMKF-TMTIR|) were calculated across trials for each condition of each subject. Group means of all subjects were also calculated for each condition. These critical event parameters were used to evaluate timing differences between the joint/segmental actions.

To examine these actions over the entire support period, additional techniques were employed. These techniques were curve correlations and velocity differences. Derrick, Bates, and Dufek (1994) introduced the curve correlation technique. Using this technique, a point-by-point Pearson product moment correlation coefficient is calculated between the corresponding data points from the angle mean ensemble curves. A high correlation always indicates similar curves and, thus, temporal similarity or proper coordination between the actions of the subtalar and the knee. A lower correlation reveals the opposite, although it is not synonymous with a lack of temporal similarity. Curve correlations were performed between the pronation and knee, knee and tibial rotation, and pronation and tibial rotation angle mean ensemble curves. Group means of all subjects and for each condition were also calculated from the curve correlation values.

For the velocity differences technique, a point-by-point difference is calculated between the corresponding data points from the angular velocity mean ensemble curves. By this technique, a new curve is generated that represents the angular velocity differences throughout the support period. Functionally, large differences between the velocities indicate antagonistic relationships and, thus, possible injurious situations. To describe the generated velocity differences curve by a single number, the mean of the absolute values of the curve were calculated. This mean captured the entire curve throughout the support period. The greater this number, the larger the velocity differences are and vice versa. Velocity differences were calculated between the pronation and knee, knee and tibial rotation, and pronation and tibial rotation angular velocity mean ensemble curves. Group means of all subjects and for each condition were also calculated from the mean absolute velocity differences (MAVD) values.

Construction of Rankings: to increase the clinical external validity of the results of this study, subjects' ranking for susceptibility to injury were constructed from both the clinical exam and the biomechanical data. The biomechanical subjects' rankings were based on the variable ability of every subject to sustain proper coordination through the changes introduced in the experiment (obstacle heights). These rankings were constructed based on the curve correlations and the velocity differences results.

The curve correlations (CC) ranking was constructed as follows. After the angular mean ensemble curves were correlated, the CC values were averaged across conditions and across test sessions for each subject. These means and their respective standard deviations were used to rank the subjects for susceptibility to injury. One subject's rank was constructed based on the means. A high mean indicated high values for CC and similar angle curves across conditions; interpreted as maintaining timing between the corresponding actions throughout the stance period. A low mean was presumed to be the opposite. The subject with the highest mean was considered as least likely to sustain an injury under varied running conditions and is assigned a rank of one. The subjects were also ranked on the standard deviations. For this ranking, a low standard deviation indicated small variations among the CC values across conditions and little dissimilarity between the two angle curves, whereas a large value was assumed to be the opposite. The subject with the lowest standard deviation value was considered as least likely to sustain an injury under varied running conditions and is assigned a rank of one. The subjects' scores from the two rankings (means and standard deviations) were then added and the sums were used to construct a final CC ranking. For this ranking, the subject with the lowest score was assigned a rank of one and vice versa.

The velocity differences (VD) ranking was constructed as follows. The MAVD values were averaged across conditions for each subject. These means and standard deviations were used to rank the subjects for susceptibility to injury. One subject's rank was constructed based on the means. A low mean value indicated smaller differences between the two angular velocities across conditions, whereas a large value was presumed to be the opposite. Large velocity differences were interpreted as antagonistic relationships and, thus, as lack of coordination. The subject with the lowest mean was considered as least likely to sustain an injury under varied running conditions and assigned a rank of one, as this subject was assumed to maintain coordination. The subjects were also ranked on the standard deviations. For this ranking, a low standard deviation indicated small variations between the angular velocity differences across conditions, whereas a large standard deviation was presumed to be the opposite. Thus, the subject with the lowest standard deviation was considered as the least likely to sustain an injury under varied running conditions and was assigned a rank of one. The subjects' scores from the two rankings (means and standard deviations) were added and the sums were used to construct a final VD ranking. For this ranking, the subject with the lowest score was assigned a rank of one and vice versa.

Clinical Evaluation and Ranking: the prevention of running injuries has been hindered by the lack of clinical evaluation to validate biomechanical results. Thus, in an effort to increase the external validity, subjects' rankings for susceptibility to injury were constructed based on a clinical evaluation and were compared with the biomechanical results. A licensed physical therapist and sports medicine specialist (Jason Paladino, MPT, ATC) conducted the clinical evaluation. The clinician ranked the subjects on susceptibility to injury based on injury history (IH), running experience (RE), and clinical examination (CE). The subject's rankings were based on the clinical expertise of Mr.

Paladino. This method is based on the work of Stergiou, et al. (1999) and is otherwise, completely subjective. However, the evaluation of patients in the clinical setting is currently based on the clinical expert's self-perception of a symptom or symptoms importance to injury, therefore, it seems reasonable to attempt to validate the biomechanical results with the evaluation of a clinical expert. Mr. Paladino, a highly respected physical therapist and certified athletic trainer who specializes in foot and ankle injuries, provided the clinical evaluation.

Statistical Analysis: One-way repeated measures ANOVA (condition by subjects) were performed on the subject means for each obstacle height condition GRIF, TMP, TMKF, TMTIR, |TMP-TMKF|, |TMP-TMTIR|, |TMKF-TMTIR|, and MAVD. In tests that resulted in a significant F ratio ($p < 0.05$), a Tukey multiple comparison test was performed to identify the location of the significant differences. Pearson product moment correlation coefficients were calculated between the GRIF and MP, MKF, MTIR, TMP, TMKF, TMTIR, |TMP-TMKF|, |TMP-TMTIR|, |TMKF-TMTIR|, and MAVD to indicate linear relationships. In addition, Spearman rank order correlations were used to compare the clinical subjects' rankings for susceptibility to injury with biomechanical subjects' rankings. All statistical tests were performed at the 0.05 alpha level.

CHAPTER IV: RESULTS

The purpose of this study was to investigate the coupling mechanism between the subtalar and knee joints during running over obstacles of varying heights using a three-dimensional analysis. The following chapter presents the results of this study, subdivided into dependent variables, and subject's rankings.

Dependent Variables

Kinetics - Ground Reaction Impact forces: The ground reaction impact force (GRIF) values for all conditions were normalized for body weight by dividing the GRIF by the subject's weight in Newtons. The group results are presented in Table 1. The individual data can be found in Appendix D. Results were comparable to those of Stergiou, et al. (1999) and Scholten (1999). The 0% condition resulted in a GRIF value of 1.87 body weights for the present study compared to 1.71 for Stergiou, et al. (1999) and 1.72 for Scholten (1999). The 5 and 10% obstacle conditions resulted in values of 2.24 and 2.36, respectively, for the present study compared to 2.02 and 2.30 for Stergiou, et al. (1999) and compared to the 10% condition value of 2.12 for Scholten (1999). However, there was some discrepancy in the 12.5% condition. Scholten (1999) reported a value of 1.99 compared to 2.51 in the present study. This difference may be due to the transition state of heel-strike to forefoot strike that Scholten (1999)

Table 1. Group means and SD for all conditions for ground reaction impact forces (GRIF). SD in parenthesis. Condition means that are significantly different ($p<0.05$) are shown in superscript.

Subject	GRIF (body weights)
0% Obstacle	$1.87^{5, 7.5, 10, 12.5\%}$ (0.31)
5% Obstacle	$2.24^{0, 12.5\%}$ (0.39)
7.5% Obstacle	$2.27^{0, 12.5\%}$ (0.41)
10% Obstacle	$2.36^{0\%}$ (0.40)
12.5% Obstacle	$2.51^{0, 5, 7.5\%}$ (0.45)

reported occurring at the 15% obstacle height condition. In the present study, subjects were required to maintain a normal heel-strike landing, which was not the case in the work of Scholten (1999). The lower values reported by Scholten (1999) may be a result of some forefoot landings.

The GRIF results were found to be statistically different among conditions. The *post hoc* analysis revealed that the condition with no obstacle (0% obstacle) was statistically different from all other conditions (Table 1). The 12.5% obstacle height was also statistically different from the 5% and 7.5% obstacle. As the obstacle height increased from one condition to the next, GRIF values increased as well. This finding was consistent with that of Stergiou, et al. (1999) although it varied from that of Scholten (1999) where no such pattern was noted. The condition with no obstacle and the 12.5% obstacle produced the largest range with a difference of 0.64. Figure 6 displays an example of this increase by depicting the Fz curves from a 0% obstacle height condition and a 12.5% obstacle height condition of a representative subject. These results indicate that the experimental design regarding the increases in obstacle height was successful in augmenting ground reaction forces. This result gives support to the first hypothesis.

Kinematics: Group results for maximum pronation (MP), maximum knee flexion (MKF), and maximum tibial internal rotation (MTIR) are presented in Table 2. Individual data can be found in Appendix E. Values for MP and MKF were consistent to those reported in literature. In the present study, MP values at the 0% condition were found to be 13.92 degrees compared to a range of 8 to 16 degrees reported in literature (Areblad, et al., 1990; McClay & Manal, 1998; Bellchamber & van den Bogert, 2000; De

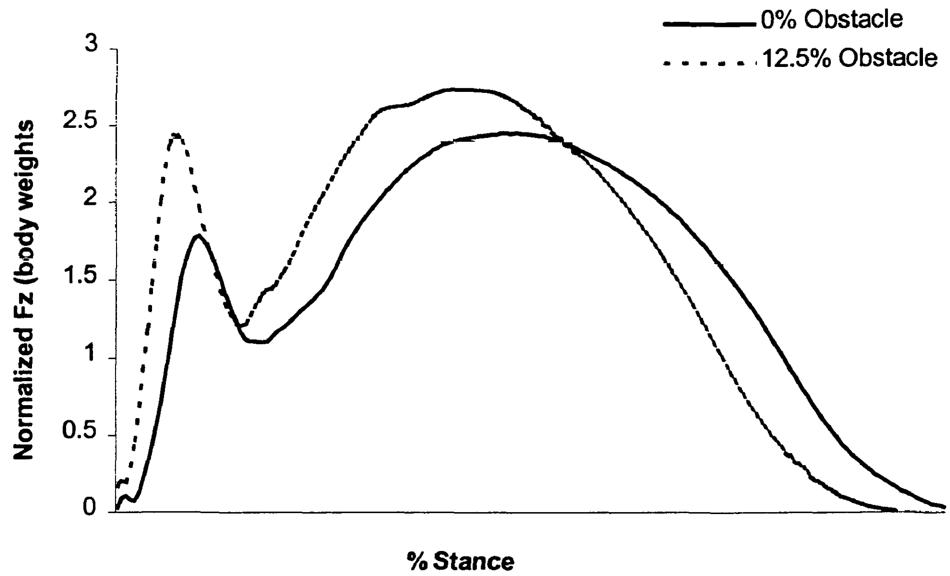

Figure 6. Fz curves from the 0% obstacle and the 12.5% obstacle of a representative subject.

Wit & De Clercq, 2000; McClay, 2000). The MKF value at 0% (37.73 degrees) was also similar to those reported in the literature (range of 37 degrees to 42 degrees) (Stergiou, 1995; De Wit & De Clercq, 2000; McClay, 2000). However, the MTIR value in the present study (2.16 degrees) was smaller than the 4.65 degrees reported by Stacoff, et al. (2000), the 5.0 degrees reported by McClay (2000), and the 12 degrees reported by Bellchamber and van den Bogert (2000). The wide range of MTIR values may be due to the high variability of the movement in the transverse plane.

Significant differences were found in the MP and the MTIR conditions. Also, in all three variables, the 0% condition had the lowest maximum value. For the MP condition, the no obstacle (0%) had less pronation than the 7.5% and the 12.5% conditions, indicating increased pronation in the obstacle conditions. For the MTIR, the 0% condition had less internal rotation than the 12.5% condition, indicating increased internal rotation in the obstacle conditions. By viewing the standard deviations (SD),

Table 2. Group means, SD, ranges (R), and GRIF correlation coefficients for maximum pronation (MP), maximum knee flexion (MKF), and maximum tibial internal rotation (MTIR). SD listed in parenthesis. Condition means that are significantly different ($p<0.05$) are shown in superscript. All values are in degrees.

Condition	MP**	R MP	MKF**	R MKF	MTIR	R MTIR**
0% Obstacle	$-13.92^{7.5,\ 12.5\%}$ (5.81)	-6.20 -24.67	37.73 (7.57)	26.56 50.61	$-2.16^{12.5\%}$ (8.02)	-16.32 6.49
5% Obstacle	-14.51 (5.63)	-7.02 -25.75	38.45 (7.60)	25.78 50.50	-1.83 (8.03)	-16.14 8.13
7.5% Obstacle	$-16.02^{0\%}$ (5.93)	-7.29 -25.95	38.24 (7.36)	27.20 50.44	-0.68 (8.50)	-15.68 9.18
10% Obstacle	-15.38 (5.38)	-7.97 -25.76	38.23 (7.34)	26.35 50.57	-1.58 (7.96)	-15.24 8.11
12.5% Obstacle	$-15.99^{0\%}$ (5.48)	-7.41 -25.69	39.34 (7.30)	28.22 50.91	$-0.48^{0\%}$ (8.08)	-15.52 9.41
Pearson r*	-0.84		0.85		0.76	

Note: range values are listed with minimum value first, maximum value second.

* correlated with ground reaction impact forces (GRIF) (Table 1).

** MP: negative = pronation, MKF: positive = flexion, MTIR: negative = external rotation.

tibial rotation was the most variable among subjects. The range for tibial rotation indicated that some subjects had maximum internal rotation occurring in an externally rotated position. These findings indicated that the transverse plane was the most variable among parameters and is consistent with that of McClay and Manal (1998). When maximum angular displacement values were correlated with GRIF results, linear relationships were observed for all variables (Table 2), although none were significant at the $p<0.05$ level. MP and MKF had the highest correlations at $r = -0.84$ (71% explained variance) and $r = 0.85$ (72% explained variance), respectively, with MTIR at $r = 0.76$ (58% explained variance). Therefore, increases in GRIF may have been related with increased maximum angular displacement for pronation, knee flexion, and tibial internal rotation.

Time variables included time to maximum pronation (TMP), time to maximum knee flexion (TMK), time to maximum tibial internal rotation (TMTIR), and the absolute differences between TMP and TMKF (|TMP-TMKF|), TMP and TMTIR (|TMP-TMTIR|), and TMKF and TMTIR (|TMKF-TMTIR|). Group results are presented in Table 3. Individual data can be found in Appendix E. Values for TMP and TMKF were comparable with those of Stergiou, et al. (1999) and De Wit and De Clercq (2000). In the present study, the 0% condition resulted in a TMP value of 37.3% of stance while Stergiou, et al. (1999) found TMP to be 44.65% of stance and De Wit and De Clercq (2000) found 41% of stance. TMP values for the 5% and 10% conditions (Table 3) were also similar to those reported by Stergiou, et al. (1999) with values of 38.64 and 41.30 percent of stance, respectively. Values for TMKF were found to be smaller for all conditions in the present study when compared to literature. Stergiou, et al. (1999) reported the 0% condition at 44.65% of stance while De Wit and De Clercq (2000)

Table 3. Group means, SD, and GRIF correlation coefficient for all conditions for time to maximum pronation (TMP), time to maximum knee flexion (TMKF), time to maximum tibial internal rotation, and absolute differences between TMP and TMKF (|TMP-TMKF|), TMP and TMTIR (|TMP-TMTIR|), and TMKF and TMTIR (|TMKF-TMTIR|). SD listed in parenthesis. Condition means that are significantly different ($p<0.05$) are shown in superscript. All values are % of stance.

Condition	TMP	TMKF	TMTIR	|TMP-TMKF|	|TMP-TMTIR|	|TMKF-TMTIR|
0% Obstacle	37.03 (2.72)	28.03 $^{12.5\%}$ (2.48)	38.41 (6.02)	9.02 $^{5\%,\ 12.5\%}$ (1.84)	5.2 (4.34)	10.9 (5.21)
5% Obstacle	37.5 (3.55)	26.57 (2.00)	38.41 (4.92)	10.93 $^{0\%}$ (2.94)	4.27 (5.41)	11.84 (5.07)
7.5% Obstacle	36.58 (5.38)	27.32 (2.73)	40.05 (5.03)	10.48 (3.45)	5.83 (6.34)	12.73 (4.87)
10% Obstacle	36.96 (3.66)	26.44 (2.64)	38.12 (5.41)	10.7 (3.15)	5.14 (4.93)	11.68 (5.36)
12.5% Obstacle	37.41 (4.00)	26.16 $^{0\%}$ (2.16)	37.98 (4.82)	11.25 $^{0\%}$ (3.24)	3.89 (4.92)	11.82 (4.67)
Pearson r*	0.22	-0.91*	-0.13	0.95*	-0.44	0.58

* correlated with ground reaction impact forces (GRIF) (Table 1). Significant at $p<0.05$.

reported 36% of stance. TMKF in the present study was found to be 28.03% of stance. Stergiou, et al. (1999) reported values of 37.58 and 37.52% of stance for the 5 and 10% obstacle conditions compared to 26.57 and 26.44 in the present study. The differences may be due to the larger standard deviations reported by Stergiou, et al. (1999). Values for |TMP-TMKF| were also smaller in the present study than those reported by Stergiou, et al. (1999). Values of 9.02, 10.93, and 10.7% of stance were found in the present study for the 0, 5, and 10% conditions, whereas Stergiou, et al. (1999) reported values of 9.86, 9.78, and 11.86% of stance, respectively.

The TMKF and |TMP-TMKF| results were found to be statistically different among conditions. The *post hoc* analysis revealed that the 0% condition was statistically different from the 12.5% obstacle condition for TMKF. For |TMP-TMKF|, it was found that the 0% obstacle condition was statistically different from the 5% and the 12.5% obstacle conditions. No significant differences were found for TMP, TMTIR, |TMP-TMTIR|, and |TMKF-TMTIR|. These results indicated that the TMKF occurred earlier (26.16% of stance) during stance in the 12.5% obstacle condition as opposed to the 0% obstacle condition, where TMKF occurred later (28.03% of stance). The earlier occurrence of TMKF in the 12.5% obstacle condition resulted in a larger |TMP-TMKF| (11.25 % of stance) when compared to the |TMP-TMKF| for the 0% obstacle condition (9.02% of stance). It should be noted that the TMP values for the two conditions were very similar values (37.03 and 37.41 % of stance) and, therefore, the TMKF was the primary contributor to the differences in |TMP-TMKF| for the 0% and 12.5% obstacle conditions. Time to maximum angular displacement values were correlated with GRIF results yielding a significant linear relationship with TMKF (r = -0.91 with 83% explained variance (Table 3). TMP and TMTIR did not reveal a distinct relationship with GRIF. Therefore, increases in GRIF were related with decreased TMKF. The absolute differences values were correlated to GRIF results as well (Table 3). The pronation and knee differences resulted in a significant linear relationship (r = 0.95 with 90% explained variance). The pronation and tibial rotation differences and also the knee and tibial rotation differences revealed poor correlations with GRIF. Therefore, increases in GRIF were related with increased timing differences between maximum pronation and maximum knee flexion.

A careful examination of the mean ensemble curves from 15% of stance to 85% of stance for pronation angles revealed that some of the pronation curves had two distinct minimums and a well-defined maximum in between (bimodal curve), whereas others displayed but one minimum (unimodal curve) (Figure 7). This is consistent with the results of Stergiou, et al. (1999). However, the overall shape of the pronation curve has some differences with those reported by Stergiou, et al. (1999), Bellchamber and van den Bogert, (2000), and McClay (2000), although none of these studies used the same methodology to model the foot. The present study does resemble that of Nigg, et al. (1993), although the presence of bimodal characteristics were absent, most likely due to over-smoothing. The first 15% of stance revealed that the foot was supinating, while the last 15% of stance displayed the foot pronating. This finding deviated from the typical 2D pronation curves in the literature and is most likely a result of the foot model used in the present study. Total foot pronation/supination was measured rather than the rearfoot

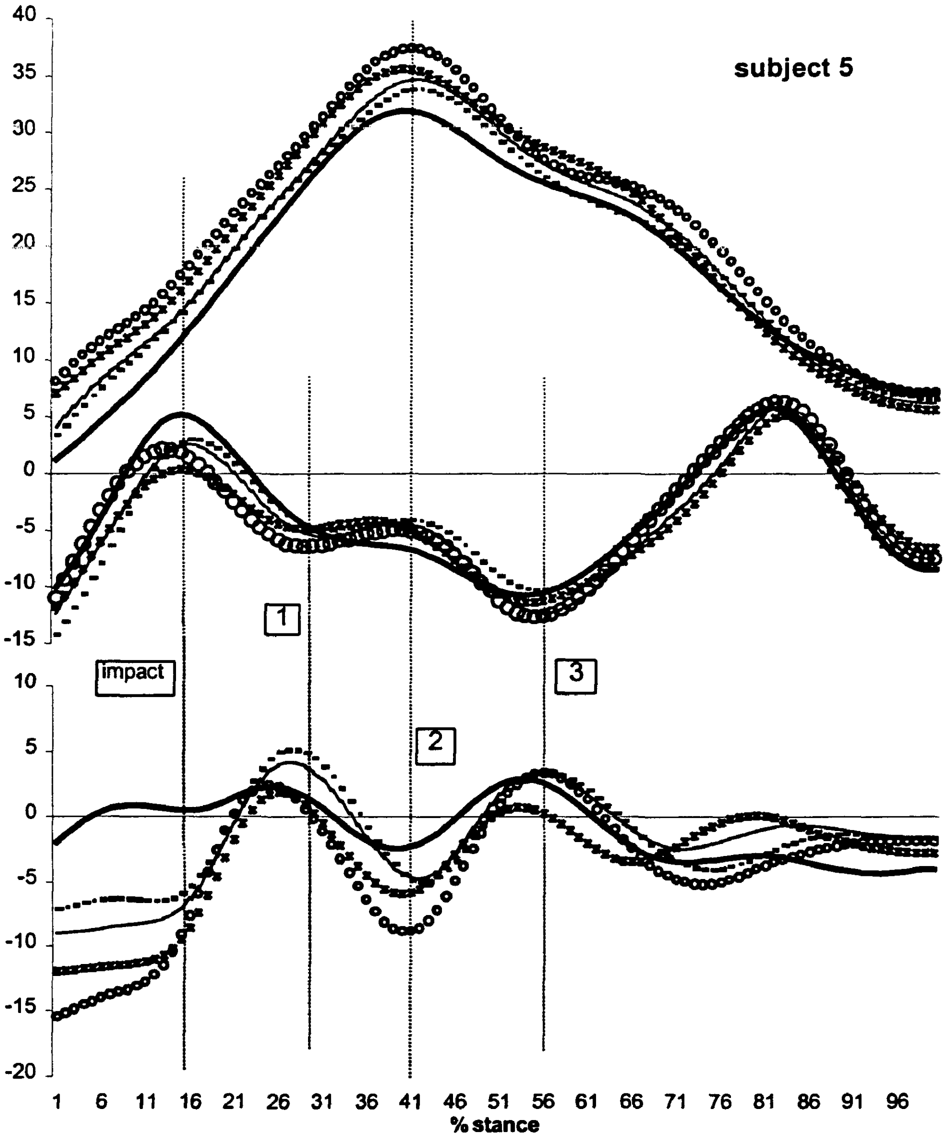

Figure 7. Pronation, knee, and tibial rotation mean ensemble curves from subject 5 for all conditions. Each curve is an ensemble average over all trials. The top group of curves are the knee angles (+ = flexion), the middle group are the pronation angles (- = pronation), and the bottom are the tibial rotation angles (+ = internal rotation). The vertical dashed lines represent points of timing sequences and are noted as impact, 1, 2, & 3. *(Conditions appear as follows: no obstacle (0%) is the thick, solid line, 5% is the coarse line, 7.5% is the dashed line, 10% is the asterisk line, and 12.5% is the cirlce line.)*

pronation/supination angle in the traditional 2D view. The foot was modeled as a rigid cone, although forefoot and rearfoot movement may act independently of one another. It is possible that at heel strike, the rearfoot became somewhat stationary while the forefoot moved or rotated medially, and foot supination developed. A similar explanation may describe the last 15% of stance in that the forefoot was mostly stationary while the rearfoot may have been moving.

The presence of bimodal and unimodal curves are depicted from a representative subject (s5) in Figure 7. It can be seen in Figure 7 that for the pronation angle curve for the condition with no obstacle (0%), there is a unimodal curve, whereas all other conditions (obstacle present) display bimodal curves. For all subjects, bimodal curves were more prevalent in the higher obstacle conditions, which corresponds to increases in GRIF. This is consistent with the findings of Stergiou, et al. (1997, 1999), which were conducted using 2D methods. The investigation of the knee angle mean ensemble curves illustrated that the presence of the obstacle and increased GRIF had little affect on the shape and timing of the curve. Lastly, the careful examination of the tibial rotation mean ensemble curves depicted the greatest amount of deviation among conditions. The overall shape of the curve was not as consistent among subjects as the pronation and knee angle curves were. Individual curves for all conditions and subjects can be found in Appendix E. Most of the tibial rotation curves displayed bimodal characteristics, with two distinct maximums for internal rotation and a minimum for external rotation located between the maximums. It can be seen in Figure 7 that the 0% condition deviated from the conditions with the obstacle present, suggesting that the presence of the obstacle and increased GRIF resulted in increased rotation of the tibia. Thus, there is a difference in timing sequences among the condition with no obstacle and when the obstacle was present, as the knee angle showed little fluctuation among conditions.

Further examination of the mean ensemble curves for pronation, knee, and tibial rotation angles yielded additional timing observations and are represented in Figure 7. Typically, the first minimum of the bimodal curve of pronation (Figure 7, line 1), which signifies a point of maximum pronation, corresponded to the first maximum of tibial rotation, which signifies a point of maximum internal rotation. It was also found that the maximum of the bimodal curve for pronation (Figure 7, line 2) (a point of max supination) occurred at approximately the same time as maximum knee flexion and also the minimum of the bimodal curve for tibial rotation (a point of max external rotation). Finally, the second minimum of the bimodal curve for pronation (Figure 7, line 3), considered to be maximum pronation, corresponded to the second maximum of tibial rotation, considered to be the point of maximum internal rotation.

Additional timing discrepancies were seen concerning the magnitude of the mean ensemble curves as well. For most subjects, the tibial rotation bimodal characteristics were pronounced in the obstacle conditions, with the higher obstacles having the greatest increase (Figure 7, Appendix E). Also, the starting position of the 0% condition tended to be of a smaller value than the obstacle conditions.

Mean Absolute Velocity Differences: To examine the actions of subtalar pronation/supination, knee flexion/extension, and tibial internal/external rotation over the entire stance period, mean absolute velocity differences (MAVD) were calculated. Group means and standard deviations for MAVD are presented in Table 4. Individual results can be found in Appendix E. MAVD values for the pronation and knee velocity differences (|P-K|) found in the present study were higher than those reported by

Table 4. Group means, SD, and GRIF correlation coefficient for all conditions for mean absolute velocity differences (MAVD). SD in parenthesis. All values are deg/s.

Condition	|P-K|	|P-T|	|K-T|
0% Obstacle	339.60 (26.92)	300.42 (93.15)	256.61 (79.04)
5% Obstacle	341.86 (38.55)	311.04 (88.57)	235.37 (47.36)
7.5% Obstacle	342.23 (47.64)	343.31 (119.99)	273.90 (76.24)
10% Obstacle	345.32 (45.45)	333.16 (88.1)	252.09 (37.85)
12.5% Obstacle	345.62 (48.28)	344.55 (118.41)	278.70 (78.04)
Pearson r*	0.92*	0.85	0.36

|P-K| = pronatin velocity minus knee velocity |P-T| = pronation velocity minus tibia velocity |K-T| = knee velocity minus tibia velocity.
* correlated with ground reaction impact forces (GRIF) (Table 1). Significant at $p<0.05$.

Stergiou, et al. (1999). They showed values of 166.58, 166.93, and 175.81 degrees per second for the 0, 5, and 10% obstacle height conditions, whereas the present study found values of 339.60, 341.86, and 345.32 degrees per second, respectively. Similar SD values were found suggesting that the differences may lie in the measurement tools.

The MAVD technique gave insight to the velocity differences throughout the stance phase. Large differences between the velocities indicated antagonistic relationships. No significant differences were found among conditions. However, for |P-K|, it can be seen that as the obstacle height increased, the velocity differences increased as well. This is analogous with the GRIF findings. Also, the largest differences for all variables occur in the 12.5% obstacle. Thus, MAVD for |P-K| exhibited a significant linear relationship with the GRIF results ($r = 0.92$ with 86% explained variance), indicating that increases in GRIF were related with greater slope differences and augmented the timing discrepancies between foot and the knee. Likewise, the |P-T| yielded a similar result ($r = 0.85$ with 72% explained variance), although not significant. These findings added support to the second hypothesis. Another interesting observation was that the |P-K| values were greater than the pronation angle and tibia angle differences |P-T| values, which were greater than the knee angle and tibia angle differences |K-T| values for all conditions except the 7.5% obstacle condition, where the |P-K| (342.23 deg/s) and |P-T| (343.31 deg/s) values were very similar.

Curve Correlations: Another technique employed to examine the entire stance period was the curve correlations (CC) technique (Derrick, et al., 1994). Group means and standard deviations for CC are presented in Table 5. By this technique, high correlations indicated similar temporal characteristics, which represents proper coordination between the ankle and knee. However, low correlations do not necessarily correspond to a lack of temporal similarity. The CC values found in the present study for the pronation angle with the knee angle (P-K) were smaller than those reported by Stergiou, et al. (1999). Stergiou, et al. (1999) reported values of 0.83, 0.79, and 0.75 for the 0, 5, and 10% obstacle height conditions compared to 0.58, 0.49, and 0.53 in the present study. The smaller correlation values are most likely due to the slight differences in the pronation curves during the first 15% and the last 15% of stance. The (P-K) and the pronation angle with tibia angle (P-T) resulted in negative correlations, indicating an inverse relationship.

Table 5. Group means and SD for curve correlations between the mean ensemble curves for all conditions. SD in parenthesis.

Condition	P-K	P-T	K-T
0% Obstacle	0.58 (0.21)	0.18 (0.32)	0.14 (0.41)
5% Obstacle	0.49 (0.26)	0.17 (0.37)	0.25 (0.41)
7.5% Obstacle	0.49 (0.25)	0.17 (0.36)	0.19 (0.30)
10% Obstacle	0.53 (0.20)	0.17 (0.35)	0.23 (0.37)
12.5% Obstacle	0.50 (0.24)	0.20 (0.39)	0.16 (0.37)

|P-K| = pronatin angle with knee angle |P-T| = pronation angle with tibia angle, |K-T| = knee angle with tibia angle.

However, the absolute value was taken for the respective correlation coefficients, as the strength of the relationship was of concern. The P-K angles showed a moderate relationship and the P-T and K-T angles resulted in poor relationships. For the P-K angles, the 0% obstacle had the highest correlation, as the obstacle conditions produced lower values. No such pattern was seen in the P-T and K-T values.

Rankings

To increase the clinical external validity of the results of this study, subjects' ranking for susceptibility to injury was constructed from both the biomechanical data and the clinical exam. The biomechanical subjects' rankings were based on the variable ability of every subject to sustain proper coordination through the changes introduced by the obstacle heights. Proper coordination was defined as the ability of a subject to maintain time matching and sequencing between the actions of the subtalar joint and the knee joint. The clinical subjects' rankings were based on the clinical exam performed by a licensed physical therapist and sport medicine specialist, who specializes in foot and ankle mechanics.

Biomechanical subjects' rankings were constructed based on the curve correlations (CC) method and the mean absolute velocity differences (MAVD) method. Individual subject rankings for CC and MAVD are presented in Tables 6 and 7, respectively. The means and standard deviations were used to rank the subjects for susceptibility to injury. For CC, subjects were ranked for pronation and knee curves (P-K), pronation and tibia curves (P-T), and knee and tibia curves (K-T). A final rank was established based on the previous three ranks. For (MAVD), subjects were ranked for pronation and knee velocity differences (P-K), pronation and tibia velocity differences (P-T), and knee and tibia velocity differences (K-T). A final rank was also established based on the previous three ranks.

Variables used to construct the clinical rankings included injury history (IH), running experience (RE), and clinical examination (CE). Individual subject rankings are presented in Table 8. The final ranking was a combination of the three variables and each variable's weight. Weights were applied to each variable based on the work of Stergiou, et al. (1999). Each of the three variables were given a weight or scale factor based on the clinician's belief as to the importance of a variable to the susceptibility to injury. Injury history was believed to play the greatest role in determining a subject's susceptibility to injury and, thus, the subject's rank was multiplied by a weight of 6. The clinical exam was second and given a weight of 3 while running experience was last and given a weight of 1.

Spearman rank order correlation coefficients were determined between the CC rankings and clinical rankings, as well as between the MAVD rankings and the clinical rankings. Subject results can be found in Table 9. No correlations were found to be significant between the clinical rankings and either biomechanical rankings.

Table 6. Subject curve correlation (CC) means and SD across conditions for each subject. The CC and SD values were used to create rankings for susceptibility to injury.

	Pronation and Knee curves (P-K)							**Knee and Tibia curves** (K-T)					
Subject	**CC**	**Rank**	**SD**	**Rank**	**Sum**	**Final**	**Subject**	**CC**	**Rank**	**SD**	**Rank**	**Sum**	**Final**
3	-0.47	7	0.11	6	13	7	3	0.24	7	0.13	7	14	8
4	-0.83	1	0.02	2	3	1	4	0.59	2	0.09	4	6	2
5	-0.30	8	0.13	10	18	9	5	0.35	5	0.09	4	9	4
6	-0.72	2	0.01	1	3	1	6	0.44	4	0.14	8	12	5
8	-0.61	4	0.06	4	8	4	8	0.09	10	0.11	6	16	9
9	-0.70	3	0.03	3	6	3	9	0.13	9	0.07	3	12	5
10	-0.29	9	0.08	6	15	8	10	-0.60	1	0.04	2	3	1
11	-0.08	10	0.10	8	18	9	11	0.35	5	0.03	1	6	2
13	-0.50	6	0.07	5	11	5	13	-0.19	8	0.19	10	18	10
14	-0.61	4	0.09	7	11	5	14	0.53	3	0.16	9	12	5

	Pronation and Tibia curves (P-T)							**Overall Rank**				
Subject	**CC**	**Rank**	**SD**	**Rank**	**Sum**	**Final**	**Subject**	**P-K**	**P-T**	**K-T**	**Sum**	**Overall Rank**
3	0.06	10	0.11	7	17	9	3	7	9	8	24	10
4	-0.47	4	0.18	10	14	7	4	1	7	2	10	2
5	-0.20	7	0.12	8	15	8	5	9	8	4	21	7
6	-0.48	3	0.06	3	6	3	6	1	3	5	9	1
8	-0.54	2	0.04	1	3	1	8	4	1	9	14	5
9	0.21	6	0.04	1	7	4	9	3	4	5	12	3
10	0.23	5	0.08	6	11	5	10	8	5	1	14	5
11	-0.68	1	0.06	3	4	2	11	9	2	2	13	4
13	0.20	7	0.07	5	12	6	13	5	6	10	21	9
14	-0.15	9	0.13	9	18	10	14	5	10	5	20	8

Note: a score of 1 corresponds to the subject that is least likely to sustain an injury, whereas a score of 10 is most likely to sustain an injury.

Table 7. Subject mean absolute velocity differences (MAVD) and SD across conditions. The MAVD and SD values were used to create rankings for susceptibility to injury.

	Pronation and Knee differences (P-K)							**Knee and Tibia differences** (K-T)					
Subject	**MAVD**	**Rank**	**SD**	**Rank**	**Sum**	**Final**	**Subject**	**MAVD**	**Rank**	**SD**	**Rank**	**Sum**	**Final**
3	283.11	2	13.40	7	9	4	3	210.00	3	6.45	1	4	1
4	358.18	5	19.21	10	15	8	4	251.94	4	63.72	8	12	6
5	317.40	3	8.81	2	5	1	5	268.07	7	32.60	7	14	8
6	365.47	6	4.17	1	7	2	6	371.85	10	88.57	10	20	10
8	385.61	10	10.94	4	14	7	8	282.68	8	15.68	2	10	4
9	326.79	4	9.54	3	7	2	9	198.96	2	22.25	4	6	2
10	365.88	7	11.45	5	12	6	10	266.85	6	64.81	9	15	9
11	379.56	9	11.95	6	15	8	11	291.26	9	19.25	3	12	6
13	373.79	8	14.35	8	16	10	13	265.28	5	25.74	5	10	4
14	273.50	1	18.37	9	10	5	14	186.45	1	29.64	6	7	3

	Pronation and Tibia differences (P-T)							**Overall Rank**				
Subject	**MAVD**	**Rank**	**SD**	**Rank**	**Sum**	**Final**	**Subject**	**P-K**	**P-T**	**K-T**	**Sum**	**Overall Rank**
3	229.22	1	25.51	5	6	3	3	4	3	1	8	2
4	360.46	7	91.54	10	17	9	4	8	9	6	23	10
5	359.83	6	61.27	9	15	6	5	1	6	8	15	4
6	516.57	10	47.54	8	18	10	6	2	10	10	22	9
8	380.90	8	45.89	7	15	6	8	7	6	4	17	6
9	232.82	3	11.31	2	5	1	9	2	1	2	5	1
10	302.81	5	11.09	1	6	3	10	6	3	9	18	7
11	407.58	9	34.16	6	15	6	11	8	6	6	20	8
13	230.98	2	15.38	3	5	1	13	10	1	4	15	4
14	243.79	4	18.46	4	8	5	14	5	5	3	13	3

Note: a score of 1 corresponds to the subject that is least likely to sustain an injury, whereas a score of 10 is most likely to sustain an injury.

Table 8. Subject rankings for the clinical exam regarding susceptibility to injury. Separate rankings were created for injury history (IH), the clinical exam (CE), and running experience (RE).

Subject	IH	CE	RE	Sum	Clinical Rank
3	9 x *6*	3 x *3*	3 x *1*	66	6
4	4 x *6*	7 x *3*	9 x *1*	54	4
5	2 x *6*	4 x *3*	4 x *1*	28	3
6	5 x *6*	10 x *3*	6 x *1*	66	6
8	3 x *6*	1 x *3*	1 x *1*	22	2
9	8 x *6*	5 x *3*	8 x *1*	71	8
10	10 x *6*	9 x *3*	2 x *1*	89	10
11	6 x *6*	6 x *3*	7 x *1*	61	5
13	1 x *6*	2 x *3*	5 x *1*	17	1
14	7 x *6*	8 x *3*	10 x *1*	76	9

Note: the criteria were weighted as for their importance to predict injuries. Factors of 6, 3, and 1 were assigned for IH, CE, and RE, respectively, and appear in italics.

Table 9. Subject clinical rankings, curve correlation (CC) rankings, and mean absolute velocity differences (MAVD) rankings. All rankings for CC and MAVD were correlated with the clinical ranking.

Subject	Clinical Rank	CC	CC P-K	CC K-T	CC P-T	MAVD	MAVD P-K	MAVD K-T	MAVD P-T
3	6	10	4	1	3	2	7	8	9
4	4	2	8	6	9	10	1	2	7
5	3	7	1	8	6	4	9	4	8
6	6	1	2	10	10	9	1	5	3
8	2	5	7	4	6	6	4	9	1
9	8	3	2	2	1	1	3	5	4
10	10	5	6	9	3	7	8	1	5
11	5	4	8	6	6	8	9	2	2
13	1	9	10	4	1	4	5	10	6
14	9	8	5	3	5	3	5	5	10
Spearman r*		-0.13	-0.35	0.05	-0.15	-0.15	0.08	-0.55	0.22

Note: P-K is pronation and knee, K-T is knee and tibia, and P-T is pronation and tibia.

*Spearman Rank Order correlation coefficient.

CHAPTER V: DISCUSSION

Previous research has investigated the mechanisms by which runners sustain chronic, overuse type injuries. However, little progress has been made in understanding the relationship between anatomic abnormalities and abnormal biomechanics of the lower extremities as they relate to specific running injuries. This study examined the coupling mechanism between the subtalar and knee joints during running over obstacles of varying heights using a three-dimensional analysis.

The results of this study indicated that ground reaction impact forces (GRIF) increased with increases in obstacle heights (Table 1), as stated in the first hypothesis. With the increases in obstacle height and GRIF, the time to maximum knee flexion (TMKF) significantly decreased between the condition with no obstacle (0%) and the highest obstacle condition (12.5%). The time it took to reach maximum knee flexion was slowest in the 0% condition and fastest in the 12.5% condition. This finding is in contrast to that of Stergiou, et al. (1999) where no differences were found. A possible explanation for this occurrence is that the increased height required to clear the largest obstacle may have caused the subject to land with increased knee flexion, thus aiding in the faster time to MKF, although this is purely conjecture and would need further investigation. Also for MKF, there was a direct linear relationship with the GRIF results, although there were no significant differences found between conditions for MKF, suggesting that increases in GRIF were related with increases in MKF (Table 2). Thus, with increases in GRIF, increasing knee flexion may have played a minor role in absorbing shock.

Pronation also serves as a shock absorbing mechanism. Significant differences were found for MP between the 0% condition and the 7.5 and 12.5% conditions (Table 2). There was also an indirect relationship with the GRIF results, which suggested that increases in GRIF were related with increases in MP to aid in shock attenuation (Table 2). MTIR resulted in similar findings to MP in that significant differences were found between the 0% and 12.5% conditions and there was a strong, direct linear relationship with the GRIF results. This also implied that increases in GRIF were related with increases in maximum internal rotation. Since there is a functional link between pronation and tibial internal rotation, this result was expected. The time to maximum pronation (TMP) and time to maximum tibial internal rotation (TMTIR) both consistently occurred later in stance than did TMKF throughout all conditions (Table 3). This suggested that shock attenuation was still occurring through pronation, and therefore tibial internal rotation, after maximum knee flexion was achieved.

Further investigation into these timing differences revealed that the decrease in TMKF resulted in increased values of the absolute differences between TMP and TMKF (|TMP-TMKF|). The 0% condition was significantly different from the 5% and 12.5% conditions with the smallest time difference occurring in the 0% condition and the largest time difference occurring in the 12.5% condition (Table 3). This was in contrast to the findings of Stergiou, et al. (1999) where no significant differences were found. Also, the |TMP-TMKF| value exhibited a strong, direct relationship with the GRIF values. These findings suggested that with increased GRIF, there was increased asynchrony between the foot and knee. The absolute differences between TMKF and TMTIR (|TMKF-

TMTIR|) and between TMP and TMTIR (|TMP-TMTIR|) did not reveal significant differences among conditions. A moderate, direct relationship was found with GRIF results for |TMKF-TMTIR| and a moderate, indirect relationship was found for |TMP-TMTIR|. The absolute differences between |TMP-TMTIR| were about half as much as those that involved the knee, which was to be expected as TMP and TMTIR occurred at very similar times.

Functionally, the timing differences of the knee with the foot and tibia indicated that the knee began its reversal from flexion to extension before the foot and tibia began their reversals of pronation to supination and internal rotation to external rotation, respectively. Furthermore, this may have indicated that the proximal end of the tibia underwent external rotation (due to knee extension) before the distal end underwent external rotation (due to the foot remaining in pronation) possibly resulting in abnormal torsional loads. Many of the most common overuse injuries in running are believed to be associated with the transfer of movement in the frontal plane (eversion) to transverse plane movement at the knee (tibial rotation) (Stergiou, et al., 1999; McClay, 2000). The above statement is strengthened with the previously mentioned results as abnormal torsional loads occurring in the tibia in the form of internal and external rotation (transverse plane movement) may be related with common overuse injuries.

Further exploration of the effect of obstacle heights and increased GRIF on the asynchrony of the knee with the foot and tibia can be seen in the mean ensemble curves between 15% and 85% of stance (Figure 7 and Appendix E, Figure E1). The presentation of the obstacle appeared to have very little affect on the general shape of the knee angle curve throughout conditions. However, the presentation of the obstacle resulted in the formation of bimodal curves between 15% and 85% of stance for pronation (one maximum between two minimums) (Figure 7, lines 1-3 of pronation curves) in most subjects and appeared to augment the bimodality in the tibial rotation curves (one minimum between two maximums) (Figure 7, lines 1-3 of tibial rotation curves) for most subjects. For both pronation and tibial rotation, the shape of the 0% condition curve appeared to deviate from the four obstacle conditions, which held relatively similar shapes for most subjects. In all subjects' pronation curves, the 0% condition had no or very little bimodal characteristics. However, the presence of the obstacle resulted in bimodal characteristics in the majority of subjects. The appearance of the maximum between the two minimums in the obstacle conditions reflects a point of maximum supination (Figure 7, line 2 of pronation curves). This point consistently occurred at about the same percent of stance as the point of maximum knee flexion. Investigation of the tibial rotation curves resulted in similar findings in that the 0% condition deviated from the four obstacle conditions, which held relatively similar shapes. All subjects displayed bimodal characteristics (Figure 7, lines 1-3 of tibial rotation curves) for the 0% condition, with the obstacle condition curves appearing to be amplified compared to the 0% curve. Functionally, the point of MKF appeared to consistently occur at the bimodal maximum supination point (maximum surrounded by two minimums) of the pronation curve and also at the bimodal maximum external rotation point (minimum surrounded by two maximums) of the tibial rotation curve (Figure 7, line 2 all curves). Also, the first minimum of the bimodal pronation curve (a point of max pronation) appeared to consistently occur at the point of the first maximum of the bimodal tibial rotation curve (a point of max internal rotation) (Figure 7, line 1). These timing occurrences suggested that, with the presence of the obstacle, there were periods of synchrony, and asynchrony among the knee with the foot, and tibia. Viewing Figure 7, from the impact line to line 1, the knee, foot, and tibia were synchronous in the actions of flexion, pronation, and tibial internal rotation, respectively. From line 1 to line 2, the knee was still flexing as the foot was supinating and the tibia was externally rotating; asynchrony occurs. Once the knee reached maximum flexion at line 2 and reversed to extension, the foot and ankle reversed to pronation and tibial internal rotation from line 2 to line 3; asynchrony continued. Once past line 3, the three actions were mostly synchronous. The above results partially supported the second hypothesis that increased GRIF would produce increased asynchrony in the actions of the knee and subtalar joints.

The increase in obstacle heights and GRIF also affected the mean absolute velocity differences (MAVD), which was used to examine the entire stance period. Although no significant differences were found among conditions, the MAVD values for the pronation and knee angle differences (|P-K|) and for the pronation and tibia angle differences (|P-T|) showed direct relationships with GRIF results, suggesting that increases GRIF were related with increases in velocity differences (Table 4). For all three MAVD variables, the 12.5% condition had the largest MAVD value. This suggested that with high GRIF values, there were greater differences in the velocity differences. Large differences between two velocities indicated antagonistic relationships and possible injurious situations. The |P-K| values were higher than the |P-T| values, which were higher than the knee and tibia differences (|K-T|). The above findings suggested that the foot and knee had the greatest antagonistic relationship that was augmented with increased GRIF. The knee and the tibia showed the smallest antagonistic relationship. Functionally, the large differences in velocity for the foot and knee are most likely a result of the coupling mechanism via the tibia. With the knee reaching maximum flexion sooner than the foot reaching maximum pronation and with the formation of bimodal curves of high magnitude in the obstacle conditions, there existed periods of asynchrony among the foot and knee. Theoretically, the transfer of motion occurs first with the foot in the frontal plane transferring movement to the tibia in the transverse plane. The tibia then transfers motion to the knee in the sagittal plane, all in a synchronous pattern. However, due to the knee reversing motion before the foot and due to bimodal pronation curves, the velocities did not reach the zero point at the same time. That is, the foot experienced three reversals of motion due to the bimodal curve. Therefore, with each reversal, zero velocity was achieved. The knee, however, reached zero once at the point of maximum pronation. This may then have caused large fluctuations in the velocities for the two variables. As a result, the tibia experienced different velocities at the distal and proximal ends as the knee pulled the proximal end into external rotation while the foot kept the distal end in internal rotation. Therefore, it seems logical that the two MAVD variables that involved the tibia revealed smaller velocity differences than the |P-K| variable. Based on this assumption, it is suggested that the proximal end of the tibia experienced the lesser antagonistic state and the distal end experienced the greater antagonistic state. This finding may give clarification to the mechanism for some overuse running injuries. Frederics (1996) stated that abnormal torsional loads on the tibia that are associated with many overuse running injuries are reported to occur in the posterior, medial edge of the tibia, most commonly the distal third. The posterior, medial edge may be synonymous with the point where internal rotation at the distal end and external rotation at the proximal end might converge.

The last technique used to examine the entire stance period was the curve correlations (CC) technique. By this technique, high correlations indicated similar temporal characteristics, although low correlations did not necessarily mean a lack of temporal similarity. Only the pronation angle with the knee angle (P-K) resulted in a meaningful correlation, although it was a moderate one (Table 5). The remaining two CC variables involved the tibia angle, which had the highest SD and range discrepancies for the maximum points. This may be a possible reason as to why the CC values were negligible when the tibia was involved. The 0% condition for the P-K variable was higher than the obstacle conditions, which were relatively the same. This suggested that the presence of the obstacle resulted in less similarity among the curves. The lower CC values for the obstacle conditions may have resulted from the presence of the bimodal curves.

The occurrence of bimodal curves in the pronation and tibial rotation angle curves was probably a mechanical phenomenon. Stergiou, et al. (1999) proposed that the first peak of the pronation bimodal curve was due to the foot angle. When the foot comes in contact with the ground in early stance, the tibia deviates laterally and then medially in a quick and sudden fashion around the fixed foot. This phenomenon is probably due to impact. The greater is the impact, the larger is the first peak. This may be explained as a rebound effect of the tibia due to contact with the ground. The harder the contact with the ground, the greater the rebound effect (Stergiou, et al., 1999). This may provide explanation as to why the bimodal curves of pronation and tibial rotation occurred at similar times during stance, given that the distal end of the tibia and the foot are linked mechanically. The aforementioned results assume that the change to bimodal characteristics may be an injury mechanism. Therefore, the system may try to avoid such a change. As it pertains to this study, an experienced runner may remain more coordinated and change later as impact forces increase, compared to a novice runner that may loose coordination with the slightest change in impact forces. Stergiou, et al. (1999) suggested that this might be due to better optimization of the available components (degrees of freedom) involved in the shock absorbing system.

A major problem in the prevention of running injuries is the absence of clinical evaluation to validate biomechanical results. Therefore, the present study attempted to increase the clinical external validity of the biomechanical results. Subject's rankings for susceptibility to injury were constructed based on the biomechanical results of CC and MAVD. No comparison was attempted between the absolute time differences and the clinical results, due to the discrete point evaluation nature of the time differences. As previously mentioned, the actions of the foot, knee, and tibia are continuous and dynamic. Thus, it is unlikely that the evaluation of specific points from the entire stance phase would be representative of these actions. No significant correlation was found for either the CC or MAVD rankings when correlated with the clinical rankings. This is in contrast to the findings of Stergiou, et al. (1999) who found that the MAVD of |P-K| rankings correlated highly ($r_s = 0.798$) with the clinical rankings. Possible explanations for the lack of correlations may be in the measurement tools. The biomechanical tools vary from those of Stergiou, et al. (1999) in that the present study was performed using a 3D analysis as opposed to a 2D analysis. Marker set-up and its interpretations varied, which may have yielded differences in the pronation curves, thus effecting the CC values. However, Stergiou, et al. (1999) did not find significant correlations for CC values. Also, the MAVD values for the present study were twice as high as those reported by Stergiou, et al. (1999) and may have affected the ranking of each subject. This may have been the result of Stergiou, et al. (1999) excluding the final 15% of stance in the MAVD calculation due to the high variability of the velocity differences. In the present study, the entire stance phase was used to determine the MAVD values, and thus, may have experienced higher MAVD values. Lastly, the clinical tools, as stated before, are purely subjective. The clinician that performed the rankings for Stergiou, et al. (1999) was a well-published orthopedist with 35 years of experience in treating runners. There may have been differences among the clinicians regarding their evaluation standards. However, the present study attempted to reproduce the methods for clinical evaluation as strictly as possible.

CHAPTER VI: SUMMARY

The purpose of this study was to investigate the coupling mechanism between the subtalar and knee joints during running over obstacles of varying heights using a three-dimensional analysis. For day 1, ten subjects ran over obstacles of four different heights (5, 7.5, 10, and 12.5%) based on the subject's standing height, as well as a condition with no obstacle. All subjects ran at a comfortable, self-selected pace. The obstacle was placed directly before a force platform so that the subject had to clear the obstacle with the right leg and land on the platform. For day 2, the same 10 subjects underwent a clinical exam performed by a sports medicine specialist. Kinetic (960 Hz) and kinematic (240 Hz) data were collected simultaneously via a Kistler force platform and a two camera Peak Motus video system. Ten trials per the five conditions were analyzed. The kinetic parameter analyzed was the vertical component of the ground reaction force, specifically the impact peak. Kinematic parameters analyzed included, maximums, minimums, and ranges for knee flexion/extension, tibial internal/external rotation, and subtalar pronation/supination angular displacements as well as their respective times of occurrence. In addition, the absolute differences between the times were also identified. To examine the actions over the entire support period, curve correlations and mean absolute velocity differences were utilized. Mean ensemble curves were generated from the normalized angular displacement and angular velocity data. Pearson product moment correlation coefficients were calculated between the ground reaction impact forces and the above parameters (excluding curve correlations) to determine relationships. To increase the clinical external validity of the results of the study, subjects' ranking for susceptibility to injury were constructed from both the clinical exam and the biomechanical data and were correlated using a Spearman rank order correlation coefficient. One-way repeated measures ANOVA (condition by subjects) were performed on the subject means for each obstacle height condition. A Tukey post-hoc was used whenever statistical differences were identified ($p<0.05$).

The results obtained from the analysis of the parameters evaluated are summarized as follows:

1. The presence of the obstacle resulted in increased GRIF with increasing obstacle heights.
2. Increasing GRIF resulted in greater degrees of maximum pronation and tibial internal rotation as well as reaching maximum knee flexion faster, but did not change the time to maximum pronation or tibial rotation. This resulted in increases between maximum knee flexion and maximum pronation time differences. The pronation curve transitioned from a unimodal to a bimodal curve during 15% and 85% of stance for most subjects. The bimodal tibial rotation curve between 15% and 85% of stance experienced increases in the bimodal characteristics for most subjects. Therefore, the increases in GRIF resulted in increased timing differences as the proximal end of the tibia began external rotation due to earlier knee flexion and the distal end maintained internal rotation due to unchanging pronation times. This may have resulted in the tibia experiencing abnormal torsional stresses that may have been augmented with increasing GRIF.

3. The biomechanical evaluations and the clinical evaluations of subject rankings regarding susceptibility to injury did not reveal a significant relationship.

Recommendations for Future Research

Based upon the results of this study, the following recommendations for future research are suggested:

1. Further investigation into the coupling mechanisms of the foot, knee, and tibia by different experimental designs. Specifically, to investigate further the actions of foot pronation/supination and tibial internal/external rotation using different methodology for modeling the lower extremity as well as different perturbations for increasing GRIF. It may also be of interest to investigate the coupling mechanism of the femur, knee, and tibia.
2. Calculate joint moments and joint powers to identify joint loads experienced during running.
3. Replicate the clinical ranking correlation with biomechanical results using more than one clinician to rank subjects to establish inter-rater reliability.

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Appendix A

Informed Consent
Health Questionnaire Form

IRB #: 540-00-EP

Title of the Research Study

A THREE-DIMENSIONAL ANALYSIS OF SUBTALAR AND KNEE JOINT COUPLING DURING RUNNING OVER OBSTACLES

Invitation

You are invited to participate in this research study. The information in this consent form is provided to help you decide whether to participate. If you have any questions, please do not hesitate to ask.

Why Are You Eligible?

Your are eligible to participate in this study because you are a healthy male or female recreational runner between the ages of 19 and 35 who runs at least three times per week. You may participate only if your are free from any present injury and physical impairment.

What Is the Purpose of This Study?

The purpose of this study is to investigate the coordination mechanism between the ankle and knee joints during running over obstacles of varying heights using a three-dimensional analysis.

What Does This Study Involve?

You will be asked to come to the Biomechanics Laboratory at the University of Nebraska at Omaha's HPER building to participate on two separate occasions. You will be asked to wear running style shorts and a comfortable short sleeved shirt on both occasions. Following the signing of this Informed Consent Form, you will be asked to fill out and sign a health questionnaire provided.

Session 1
The first testing session will take approximately two hours. Reflective markers will be placed on your joints (i.e. the knee). You will then be instructed to run at a self-selected comfortable pace along a 30-meter runway. This will allow you time to warm-up and establish a comfortable running pace. A marker on the floor will be used as a cue to achieve a right foot contact on the force platform.

You will be asked to look straight ahead and run through the test set-up without making a conscious effort to hit the platform. You will then be required to run at your self-selected pace over four different obstacle height trials and one trial with no obstacle. You will be required to run no more than 75 times (running trials) during the testing session and will be allowed an ample rest period between each trial. While you run at your selected pace, your running performance will be recorded by two cameras.

Session 2
The second session will take approximately one hour. You will be examined by a physical therapist. The physical therapist will administer a battery of orthopedic tests to examine lower extremity alignment and geometry. The specific tests are indicated on the attached form. None of the measures will require you to assume any position that is physically stressful. In addition to the orthopedic tests, the physical therapist will ask you about your running experience and injury history.

What Are the Possible Risks and Discomforts You Could Experience?

Possible risks and discomforts you could experience during this study include: muscle soreness, dizziness, fainting, shortness of breath, and/or reduced coordination. Other possible risks and discomforts, however highly unlikely, include: abnormal heart rhythms, abnormal blood pressure, chest pain, heart attack, stroke, and/or sudden death.

What Are the Possible Benefits to You?

Opportunity to learn about your own running characteristics and your potential for running injuries. This may be useful in shoe selection, training, and technique improvement. All data gathered during this study will be made available for review upon your request. By participating in this study, you will also gain experience in participating in an experiment.

What Are the Possible Benefits to Society?

The information in this study may enhance our understanding of lower extremity function during running. This may allow us to uncover possible mechanisms that may be signs of injury. By evaluating these signs, we may then provide information for prevention of some running injuries. Therefore, this study has the potential to benefit other professions, such as, neuroscience, orthopaedics, health practitioners, coaches, and any other discipline that has an interest in preventing foot injuries, including, shoe construction, robotics, design of artificial limbs and general research on running.

Initials

What Should You Do In Case of an Emergency?

If you have a research related injury or problem, you should immediately contact one of the investigators listed at the end of this consent form.

How Will Your Confidentiality Be Protected?

Information obtained from you in this study will be treated confidentially. The only people who will have access to your research records are the primary investigator, the secondary investigator, the Institutional Review Board (IRB), and any other person or agency required by law. The information from this study may be published in scientific journals or presented at scientific meetings but your identity will be kept strictly confidential.

What Are Your Rights as a Research Participant?

You have rights as a research participant. These rights are explained in *The Rights of Research Participants,* which you have been given. If you have any questions concerning your rights, you may contact the Institutional Review Board (IRB), telephone (402) 559-6463.

What Will Happen if You Decide Not to Participate?

You can decide not to participate in this study or you can withdraw from this study at any time. Your decision will not affect your care or your relationship with the investigators, the University of Nebraska at Omaha, the University of Nebraska Medical Center, or the Nebraska Health System (NHS). Your decision will not result in any loss of benefits to which you are entitled. If any new information develops during the course of this study that may affect your willingness to continue participating, you will be informed immediately.

Documentation of Informed Consent

YOU ARE VOLUNTARILY MAKING A DECISION WHETHER TO PARTICIPATE IN THIS RESEARCH. YOUR SIGNATURE MEANS THAT YOU HAVE READ AND UNDERSTOOD THE INFORMATION PRESENTED AND DECIDED TO PARTICIPATE. YOUR SIGNATURE ALSO MEANS THAT THE INFORMATION ON THIS CONSENT FORM HAS BEEN FULLY EXPLAINED TO YOU AND ALL YOUR QUESTIONS HAVE BEEN ANSWERED TO YOUR SATISFACTION. IF YOU THINK OF ANY

Initials ______

ADDITIONAL QUESTIONS DURING THE STUDY, YOU SHOULD CONTACT THE INVESTIGATOR(S). YOU WILL BE GIVEN A COPY OF THIS CONSENT FORM.

______________________________ ______________

SIGNATURE OF PARTICIPANT **DATE**

I CERTIFY THAT ALL THE ELEMENTS OF INFORMED CONSENT DESCRIBED ON THIS CONSENT FORM HAVE BEEN EXPLAINED FULLY TO THE PARTICIPANT. IN MY JUDGEMENT, THE PARTICIPANT IS VOLUNTARILY AND KNOWINGLY GIVING INFORMED CONSENT AND POSSESSES THE LEGAL CAPACITY TO GIVE INFORMED CONSENT TO PARTICIPATE IN THIS RESEARCH.

______________________________ ______________

SIGNATURE OF INVESTIGATOR **DATE**

Authorized Study Personnel

Primary Investigator:

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(Work) 554-3225

Secondary Investigator:

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Assistant Professor, School of HPER
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HEALTH QUESTIONNAIRE FORM

Name ______________________________ Height ______________

Address ____________________________ Weight______________

Phone ______________________________ Age ______________

Are you a smoker? Yes_____ No_____

Medical History
Check any of the following which has occurred in your medical history:

_____ 1. Heart attack

_____ 2. High blood pressure

_____ 3. Angina pectoris (chest pain)

_____ 4. ECG abnormality

_____ 5. Obesity

_____ 6. Diabetes

_____ 7. Epilepsy

_____ 8. Asthma

_____ 9. Emphysema

_____ 10. Flat feet

_____ 11. Lower back ache

_____ 12. Other_________

List and describe any condition(s) you have which may affect your ability to participate in this study. (e.g. muscle sprains, tendonitis)

____

List any medication(s) you are presently taking and the condition being treated.

____

Describe your current exercise habits.

____

Any family history of: (check if yes)

_____ Heart Disease _____ Diabetes
_____ High blood pressure _____ Stroke

I have answered the above questions to the best of my ability.

_________________________________ ____________________
Signature Date

Appendix B

Clinical Evaluation

Clinical Evaluation

1. Angle of Gait
a. In-toe R L
b. Straight R L
c. Out-toe R L

2. Genu Varum
a. Valgum
b. Straight

3. Tibial Torsion
a. None
b. Internal
c. External
d. Squinting patellae

4. Leg Varus (to floor) R L

5. Foot Type Standing
a. High arch R L
b. Low arch R L
c. Medium R L

6. Standing Pronation
a. Maximum R L
b. Add. avail R L

7. Extremity Length R L

8. Ankle Dorsiflexion
a. Knee extended R L
b. Knee flexed R L

9. Hip Rotation (prone)
a. Internal R L
b. External R L
c. Symmetrical R L

10. Foot Motions
a. Loose R L
b. Tight R L
c. Normal R L

11. Subtalar Joint
a. Loose R L
b. Normal R L
c. Restricted R L
d. Varus R L
e. Neutral R L
f. Valgus R L
g. Inversion R L
h. Eversion R L

12. Forefoot Alignment
a. Neutral R L
b. Varus R L
c. Valgus R L

13. Position 1st Ray
a. Plantarflexed R L
b. Dorsiflexed R L
c. Neutral R L

14. Motion 1st Ray
a. Normal R L
b. Mod. restricted R L
c. Restricted R L

15. 1st MPJ Motion
a. Restricted R L
b. Dorsiflexion R L
c. Painful R L
d. Tender R L

16. Toe Position
a. Straight R L
b. Contracted R L
c. Subluxed R L
d. Hall. Val. R L

17. Location callus/corns

Appendix C

Subject Demographics

Table C1. Subject demographics.

Subject	Age	Weight (kg)	Height (cm)	Sex
*1	24	63.29	165.735	F
*2	25	71.34	177.8	M
3	21	61.36	162.56	F
4	28	73.64	187.96	F
5	27	81.82	177.8	M
6	35	79.54	185.42	M
*7	25	66.82	175.26	M
8	23	77.27	180.34	M
9	25	65.9	170.18	M
10	28	66.59	177.8	F
11	22	62.73	162.56	F
*12	25	74.09	175.26	F
13	22	84.09	187.96	M
14	22	63.63	172.72	F
**Mean	25.3	71.66	176.53	5 F
**SD	4.32	8.59	9.45	5 M

* subjects were collected but not included in data analysis.

** does not include subjects 1, 2, 7, and 12.

Appendix D

Individual Subject Kinetic Parameters

Table D1. Mean and SD of Ground Reaction Impact Force (GRIF) for all conditions and subjects. All values have been normalized and are in body weights.

Subject	0% Obstacle	5% Obstacle	7.5% Obstacle	10% Obstacle	12.5% Obstacle
3	1.59	2.47	2.70	2.46	2.79
4	1.66	1.80	1.68	1.82	2.08
5	2.36	2.61	2.59	2.85	3.09
6	2.31	2.59	2.82	2.82	2.96
8	1.83	1.94	1.94	2.10	2.17
9	1.68	2.49	2.52	2.56	2.64
10	1.76	1.75	1.91	1.93	1.82
11	2.13	2.34	2.36	2.62	2.71
13	1.93	2.66	2.43	2.57	2.82
14	1.45	1.73	1.78	1.83	2.02
Mean	1.87	2.24	2.27	2.36	2.51
SD	0.31	0.39	0.41	0.40	0.45

Appendix E

Individual Subject Kinematic Parameters

Table E1. Mean, SD, and ranges of maximum pronation for all conditions and subjects. SD for subjects in parenthesis. All values are degrees.

Subject	0%	5%	7.5%	10%	12.5%
3	-8.84	-10.13	-10.85	-9.75	-10.53
	(1.18)	(2.00)	(1.64)	(1.16)	(2.14)
4	-6.20	-9.09	-14.77	-12.74	-14.78
	(1.64)	(1.98)	(1.14)	(1.43)	(1.63)
5	-10.90	-12.54	-10.52	-11.84	-13.48
	(0.83)	(2.06)	(1.25)	(0.93)	(2.61)
6	-19.30	-18.02	-22.63	-18.17	-20.10
	(2.18)	(2.29)	(1.69)	(1.83)	(2.14)
8	-18.11	-20.23	-21.50	-21.23	-21.83
	(0.71)	(1.81)	(1.69)	(2.26)	(1.56)
9	-12.57	-12.43	-13.26	-14.16	-12.90
	(0.60)	(0.54)	(0.84)	(1.22)	(0.80)
10	-13.64	-15.36	-17.21	-17.02	-17.93
	(0.80)	(1.30)	(1.89)	(1.45)	(2.00)
11	-7.91	-7.02	-7.29	-7.97	-7.41
	(1.42)	(0.95)	(0.98)	(0.99)	(1.25)
13	-24.67	-25.75	-25.95	-25.76	-25.69
	(1.40)	(1.93)	(1.33)	(0.78)	(1.02)
14	-17.03	-14.51	-16.16	-15.19	-15.23
	(1.52)	(2.94)	(2.29)	(1.32)	(1.35)
Mean	-14.48	-14.99	-16.59	-16.01	-16.59
SD	5.87	5.74	5.98	5.30	5.44
Max	-24.67	-25.75	-25.95	-25.76	-25.69
Min	-6.20	-7.02	-7.29	-7.97	-7.41

Table E2. Mean, SD, and ranges of maximum knee flexion for all conditions. SD for subjects in parenthesis. All values are degrees.

Subject	0%	5%	7.5%	10%	12.5%
3	39.18	40.54	40.17	41.20	40.02
	(1.47)	(1.44)	(1.65)	(1.11)	(2.30)
4	42.06	39.84	38.45	37.69	38.94
	(0.82)	(3.05)	(2.26)	(1.63)	(1.94)
5	32.04	34.81	33.96	35.83	37.84
	(1.15)	(1.87)	(2.00)	(1.02)	(2.62)
6	31.31	31.05	31.62	31.93	31.38
	(1.62)	(1.84)	(1.77)	(1.81)	(2.32)
8	42.48	45.08	45.88	45.64	47.58
	(2.29)	(1.52)	(1.61)	(2.07)	(1.54)
9	35.31	34.96	34.68	34.07	34.70
	(1.40)	(1.64)	(1.76)	(1.19)	(2.16)
10	31.81	35.00	33.92	33.93	36.70
	(0.92)	(1.62)	(1.67)	(1.54)	(2.50)
11	50.61	50.50	50.44	50.57	50.91
	(2.69)	(1.86)	(1.73)	(2.48)	(2.06)
13	45.91	46.95	46.05	45.08	47.10
	(1.41)	(2.19)	(1.82)	(1.27)	(1.40)
14	26.56	25.78	27.20	26.35	28.22
	(1.96)	(2.91)	(1.88)	(1.76)	(2.54)
Mean	37.73	38.45	38.24	38.23	39.34
SD	7.57	7.60	7.36	7.34	7.30
Max	50.61	50.50	50.44	50.57	50.91
Min	26.56	25.78	27.20	26.35	28.22

Table E3. Mean, SD, and ranges of maximum tibial internal rotation for all conditions and subjects. SD for subjects in parenthesis. All values are degrees.

Subject	0%	5%	7.5%	10%	12.5%
3	-16.32	-16.14	-15.68	-15.24	-15.52
	(1.37)	(0.72)	(1.03)	(0.97)	(1.08)
4	-5.59	-5.45	-5.06	-3.85	-3.75
	(1.98)	(2.26)	(2.28)	(1.95)	(1.54)
5	3.03	3.66	3.68	1.02	4.49
	(0.83)	(1.36)	(2.29)	(1.80)	(2.70)
6	6.28	4.09	7.24	1.43	5.59
	(2.94)	(4.20)	(2.79)	(3.05)	(2.10)
8	-1.61	-1.45	1.37	1.81	1.07
	(1.53)	(1.28)	(1.36)	(1.67)	(1.33)
9	0.87	1.70	1.75	2.23	3.22
	(1.84)	(1.15)	(1.55)	(1.22)	(0.86)
10	-7.87	-6.97	-5.80	-5.10	-2.17
	(0.86)	(1.23)	(2.09)	(1.90)	(4.05)
11	-12.05	-11.79	-11.46	-13.74	-12.29
	(1.46)	(1.94)	(1.41)	(1.98)	(1.28)
13	6.49	5.91	9.18	7.52	5.19
	(5.15)	(1.52)	(2.13)	(1.36)	(0.96)
14	5.19	8.13	7.94	8.11	9.41
	(0.90)	(1.60)	(1.55)	(1.98)	(1.66)
Mean	-2.16	-1.83	-0.68	-1.58	-0.48
SD	8.02	8.03	8.50	7.96	8.08
Max	6.49	8.13	9.18	8.11	9.41
Min	-16.32	-16.14	-15.68	-15.24	-15.52

Note: negative values are external rotation, positive values are internal rotation.

Table E4. Means and SD for time to maximum pronation for all conditions and subjects. SD appears in parenthesis. All values are % of stance.

Subject	0%	5%	7.5%	10%	12.5%
3	40.00	42.60	46.20	41.30	45.60
	(5.75)	(4.09)	(4.92)	(1.89)	(4.62)
4	39.50	38.80	27.70	37.20	38.10
	(2.51)	(1.55)	(5.58)	(5.49)	(1.91)
5	34.90	34.70	34.80	33.90	35.00
	(1.52)	(1.16)	(0.79)	(1.37)	(1.05)
6	37.40	38.10	35.40	39.40	37.90
	(1.17)	(1.37)	(0.84)	(1.71)	(1.29)
8	37.40	39.50	40.50	39.20	39.90
	(2.63)	(1.08)	(1.18)	(2.25)	(1.66)
9	33.00	30.30	31.30	30.90	31.10
	(1.05)	(1.49)	(1.42)	(1.20)	(1.20)
10	36.80	38.50	37.00	35.70	34.30
	(3.77)	(0.97)	(2.31)	(1.83)	(1.34)
11	34.40	33.80	33.30	32.40	35.00
	(4.03)	(2.44)	(2.21)	(3.44)	(1.63)
13	35.30	38.90	37.60	38.10	37.10
	(4.16)	(3.87)	(1.35)	(4.82)	(2.73)
14	41.60	39.80	42.00	41.50	40.10
	(3.37)	(1.55)	(1.94)	(1.65)	(2.08)
Mean	37.03	37.50	36.58	36.96	37.41
SD	2.72	3.55	5.38	3.66	3.98

Table E5. Means and SD for time to maximum knee flexion for all conditions and subjects. SD appears in parenthesis. All values are % of stance.

Subject	0%	5%	7.5%	10%	12.5%
3	28.60	27.60	27.80	27.10	27.30
	(1.65)	(1.43)	(1.40)	(1.60)	(1.64)
4	31.30	29.30	33.00	30.90	28.70
	(0.67)	(1.49)	(0.94)	(0.57)	(1.42)
5	26.40	26.30	26.20	25.30	25.90
	(1.17)	(0.82)	(1.03)	(0.95)	(1.37)
6	28.70	27.40	26.30	27.60	27.60
	(1.16)	(0.97)	(0.48)	(1.17)	(1.35)
8	28.30	27.80	30.10	28.20	27.70
	(1.42)	(2.25)	(1.37)	(1.93)	(1.64)
9	26.40	23.50	23.90	23.00	23.40
	(0.70)	(1.65)	(1.91)	(1.15)	(1.51)
10	29.10	26.10	25.20	24.10	24.00
	(1.60)	(1.20)	(1.40)	(1.20)	(0.82)
11	27.30	26.00	24.60	27.20	26.70
	(2.00)	(2.11)	(2.84)	(3.65)	(2.31)
13	22.90	23.20	27.90	22.60	22.40
	(1.10)	(1.48)	(4.98)	(0.84)	(1.26)
14	31.30	28.50	28.20	28.40	27.90
	(1.95)	(1.72)	(1.03)	(1.65)	(1.37)
Mean	28.03	26.57	27.32	26.44	26.16
SD	2.48	2.00	2.73	2.64	2.16

Table E6. Means and SD for time to maximum tibial internal rotation for all conditions and subjects. SD appears in parenthesis. All values are % of stance.

Subject	0%	5%	7.5%	10%	12.5%
3	44.30	39.00	40.20	36.70	39.40
	(5.77)	(1.33)	(1.62)	(3.50)	(1.43)
4	36.70	38.50	43.50	41.10	39.40
	(1.49)	(2.42)	(1.58)	(2.02)	(1.71)
5	34.70	34.60	34.90	33.00	35.10
	(1.06)	(1.07)	(1.10)	(1.15)	(1.10)
6	38.20	38.90	35.50	41.40	38.70
	(1.40)	(1.37)	(0.71)	(4.25)	(1.49)
8	42.60	42.60	43.30	41.70	42.10
	(0.84)	(2.01)	(1.42)	(1.25)	(0.74)
9	48.00	48.20	48.00	48.10	47.20
	(5.62)	(0.63)	(1.05)	(1.10)	(1.14)
10	27.40	38.10	37.40	36.50	34.30
	(3.24)	(1.20)	(1.17)	(1.08)	(1.16)
11	37.50	35.20	32.50	35.10	34.70
	(2.01)	(2.10)	(1.72)	(5.86)	(1.16)
13	32.80	29.60	45.60	28.60	29.70
	(2.49)	(1.71)	(8.77)	(0.70)	(2.41)
14	41.90	39.40	39.60	39.00	39.20
	(1.45)	(1.84)	(1.07)	(2.26)	(1.55)
Mean	38.41	38.41	40.05	38.12	37.98
SD	6.02	4.92	5.03	5.41	4.82

Table E7. Means and SD for the absolute time differences between max pronation and max knee flexion for all conditions and all subjects. SD appears in parenthesis. All values are in % of stance.

Subject	0%	5%	7.5%	10%	12.5%
3	11.4	15.0	18.4	14.2	18.3
	(5.02)	(3.65)	(4.74)	(2.82)	(5.12)
4	8.2	9.5	6.9	7.3	9.4
	(2.44)	(1.51)	(2.69)	(3.83)	(1.51)
5	8.5	8.4	8.6	8.6	9.1
	(0.85)	(0.84)	(0.52)	(0.84)	(1.10)
6	8.7	10.7	9.1	11.8	10.3
	(1.16)	(1.06)	(0.74)	(1.32)	(0.67)
8	9.1	11.7	10.4	11.0	12.2
	(2.18)	(2.21)	(1.51)	(0.94)	(1.23)
9	6.6	6.8	7.4	7.9	7.7
	(0.97)	(1.40)	(1.17)	(0.74)	(0.67)
10	7.7	12.4	11.8	11.6	10.3
	(3.92)	(1.17)	(2.10)	(1.58)	(1.64)
11	7.3	7.8	8.7	6.0	8.3
	(3.06)	(2.25)	(2.45)	(3.71)	(2.67)
13	12.4	15.7	9.7	15.5	14.7
	(3.37)	(3.97)	(4.55)	(4.35)	(3.23)
14	10.3	11.3	13.8	13.1	12.2
	(2.11)	(1.49)	(1.62)	(0.74)	(1.23)
Mean	9.0	10.9	10.5	10.7	11.3
SD	1.84	2.94	3.45	3.15	3.24

Table E8. Means and SD for the absolute time differences between max knee flexion and max tibial internal rotation for all conditions and all subjects. SD appears in parenthesis. All values are in % of stance.

Subject	0%	5%	7.5%	10%	12.5%
3	15.7	11.4	12.4	9.6	12.1
	(6.02)	(1.51)	(1.78)	(2.50)	(1.52)
4	5.4	9.2	10.5	10.2	10.7
	(1.35)	(3.61)	(1.84)	(2.20)	(1.42)
5	8.3	8.3	8.7	7.7	9.2
	(0.48)	(0.67)	(0.82)	(0.67)	(1.32)
6	9.5	11.5	9.2	13.8	11.1
	(0.71)	(1.65)	(0.63)	(3.68)	(1.29)
8	14.3	14.8	13.2	13.5	14.4
	(1.42)	(2.94)	(1.62)	(1.51)	(1.58)
9	21.6	24.7	24.1	25.[illegible]	23.8
	(5.91)	(1.83)	(1.91)	(0.88)	(1.03)
10	3.5	12.0	12.2	12.4	10.3
	(2.12)	(0.00)	(1.62)	(1.07)	(1.16)
11	10.2	9.2	7.9	7.9	8.0
	(1.87)	(2.53)	(1.73)	(4.12)	(2.94)
13	9.9	6.4	17.7	6.0	7.3
	(2.92)	(0.84)	(6.17)	(0.67)	(2.71)
14	10.6	10.9	11.4	10.6	11.3
	(2.07)	(1.45)	(1.07)	(1.17)	(0.67)
Mean	10.9	11.8	12.7	11.7	11.8
SD	5.21	5.07	4.87	5.36	4.67

Table E9. Means and SD for the absolute time differences between max pronation and max tibial internal rotation for all conditions and all subjects. SD appears in parenthesis. All values are in % of stance.

Subject	0%	5%	7.5%	10%	12.5%
3	7.1	3.8	6.6	4.6	6.4
	(4.77)	(2.90)	(4.22)	(4.06)	(3.75)
4	3.2	2.5	15.8	5.7	1.7
	(2.04)	(1.50)	(6.07)	(4.08)	(1.34)
5	0.8	0.7	0.5	0.9	0.7
	(0.79)	(0.67)	(0.53)	(0.74)	(0.48)
6	1.0	1.2	0.5	2.8	1.0
	(1.05)	(1.81)	(0.53)	(2.30)	(1.25)
8	5.2	3.1	2.8	2.7	2.4
	(2.57)	(2.60)	(1.40)	(1.57)	(1.35)
9	15.0	17.9	16.7	17.2	16.1
	(5.60)	(1.45)	(1.57)	(0.92)	(0.88)
10	9.4	0.8	1.4	1.0	0.6
	(3.69)	(0.92)	(1.35)	(1.56)	(0.70)
11	3.7	2.0	1.2	4.5	1.5
	(3.71)	(2.98)	(1.62)	(5.54)	(1.43)
13	4.3	9.3	10.4	9.5	7.4
	(4.08)	(4.40)	(5.48)	(4.84)	(4.43)
14	2.3	1.4	2.4	2.5	1.1
	(2.16)	(0.84)	(1.43)	(1.43)	(0.99)
Mean	5.2	4.3	5.8	5.1	3.9
SD	4.34	5.41	6.34	4.93	4.92

Table E10. Mean absolute velocity differences between pronation and knee flexion for all conditions and all subjects. All values are degrees per second.

Subject	0%	5%	7.5%	10%	12.5%
3	298.37	289.28	276.35	287.90	263.66
4	331.94	351.06	365.45	358.10	384.34
5	320.36	322.18	304.63	312.74	327.08
6	367.82	359.42	367.34	369.72	363.03
8	374.32	376.79	382.69	396.98	397.28
9	331.65	318.61	331.72	337.12	314.85
10	347.39	369.22	377.84	370.87	364.08
11	364.47	388.02	393.77	371.22	380.31
13	356.85	368.97	365.68	389.61	387.85
14	302.84	275.10	256.89	259.00	273.70
Mean	339.60	341.86	342.23	345.32	345.62
SD	26.92	38.55	47.64	45.45	48.28

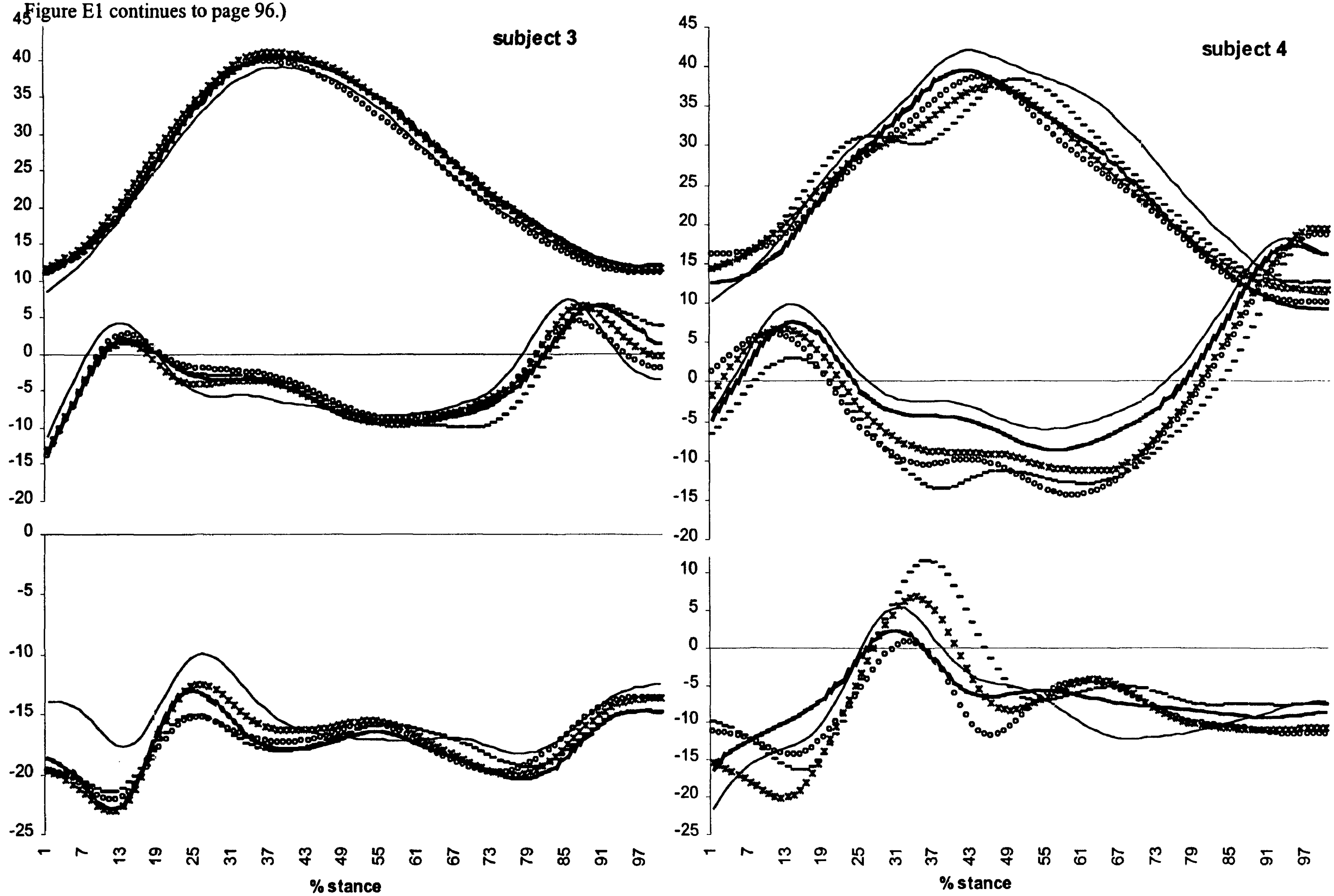
Table E11. Mean absolute velocity differences between knee flexion and tibial internal rotation for all conditions and all subjects. All values are degrees per second.

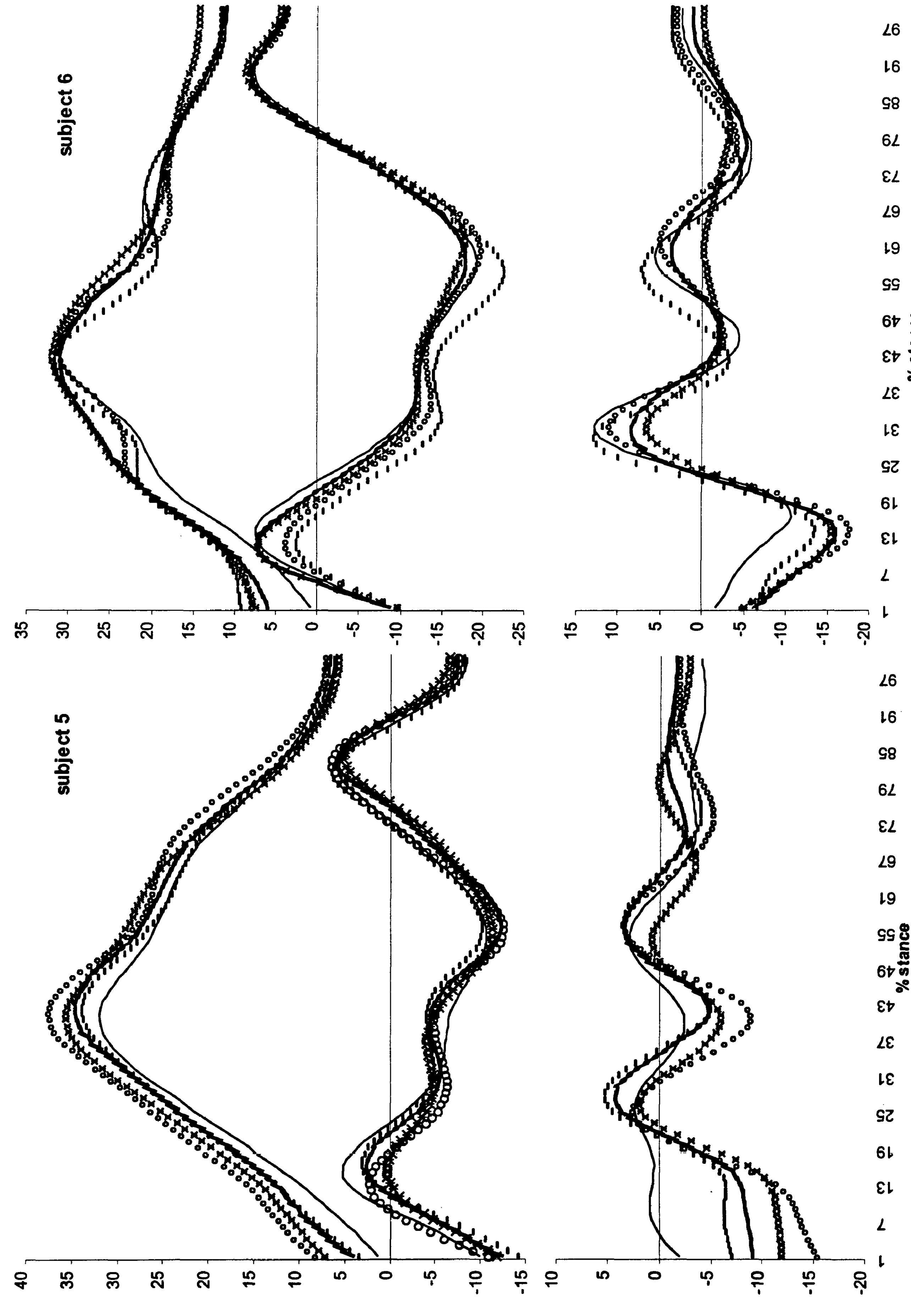
Subject	0%	5%	7.5%	10%	12.5%
3	248.72	238.42	197.87	254.55	206.55
4	260.49	264.03	452.17	425.68	399.93
5	281.10	376.50	397.76	313.55	430.25
6	527.62	481.35	558.47	454.02	561.39
8	344.77	331.44	387.43	447.86	393.02
9	221.37	223.59	242.73	246.54	229.85
10	295.87	297.78	307.06	320.28	293.07
11	361.94	416.17	431.50	383.71	444.57
13	220.78	223.19	234.47	256.45	220.00
14	241.50	257.93	223.68	228.96	266.87
Mean	300.42	311.04	343.31	333.16	344.55
SD	93.15	88.57	119.99	88.10	118.41

Table E12. Mean absolute velocity differences between pronation and tibial internal rotation for all conditions and all subjects. All values are degrees per second.

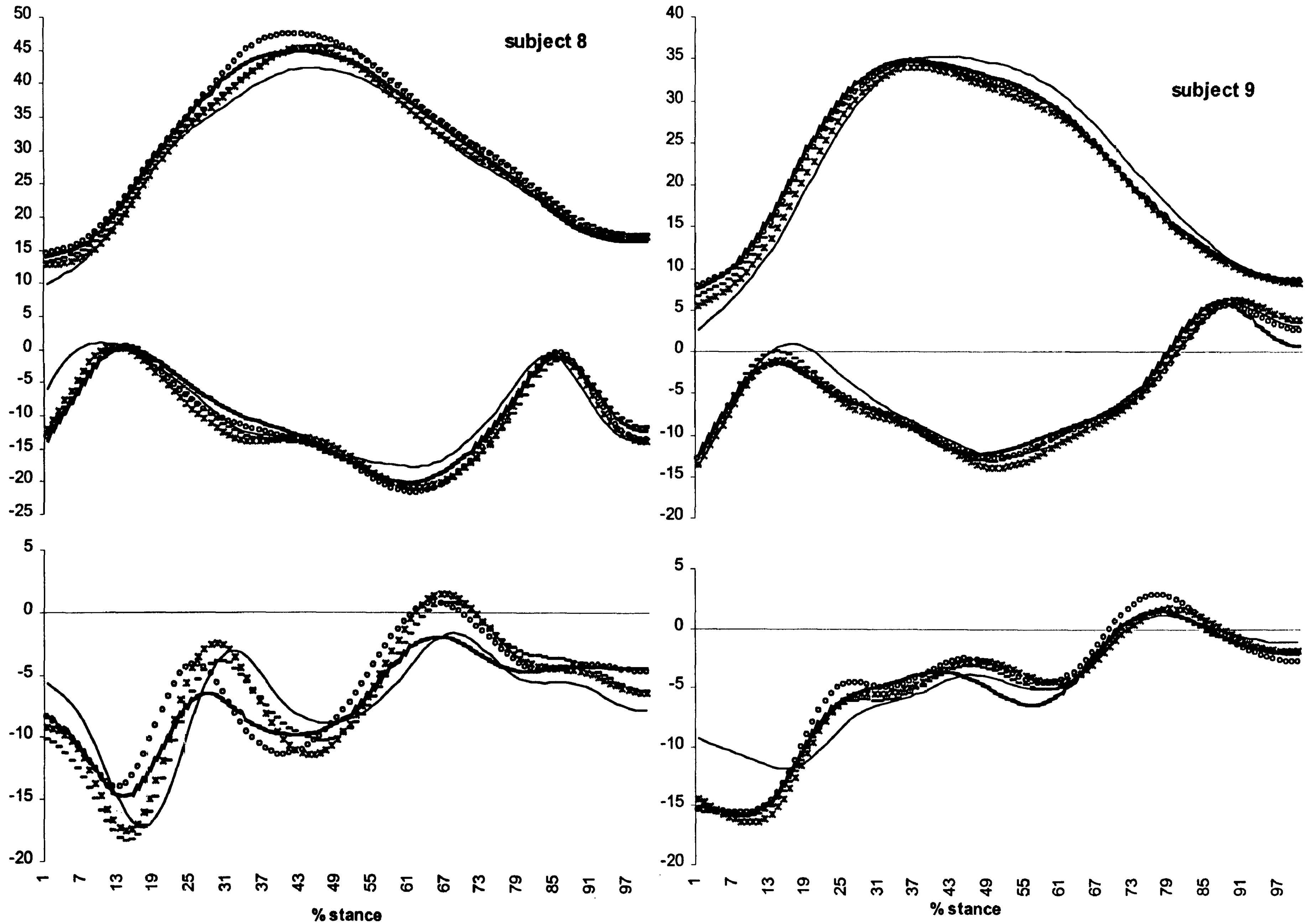
Subject	0%	5%	7.5%	10%	12.5%
3	219.01	212.94	210.18	205.26	202.63
4	219.68	188.78	354.11	267.93	229.19
5	230.97	262.52	270.15	256.60	320.11
6	435.03	312.41	448.60	245.81	417.41
8	290.19	258.72	276.94	299.87	287.69
9	236.45	183.00	194.41	199.26	181.65
10	191.44	249.36	247.00	278.18	368.27
11	285.66	274.88	276.87	296.91	321.99
13	308.78	250.40	245.04	267.84	254.34
14	148.84	160.71	215.73	203.29	203.69
Mean	256.61	235.37	273.90	252.09	278.70
SD	79.04	47.36	76.24	37.85	78.04

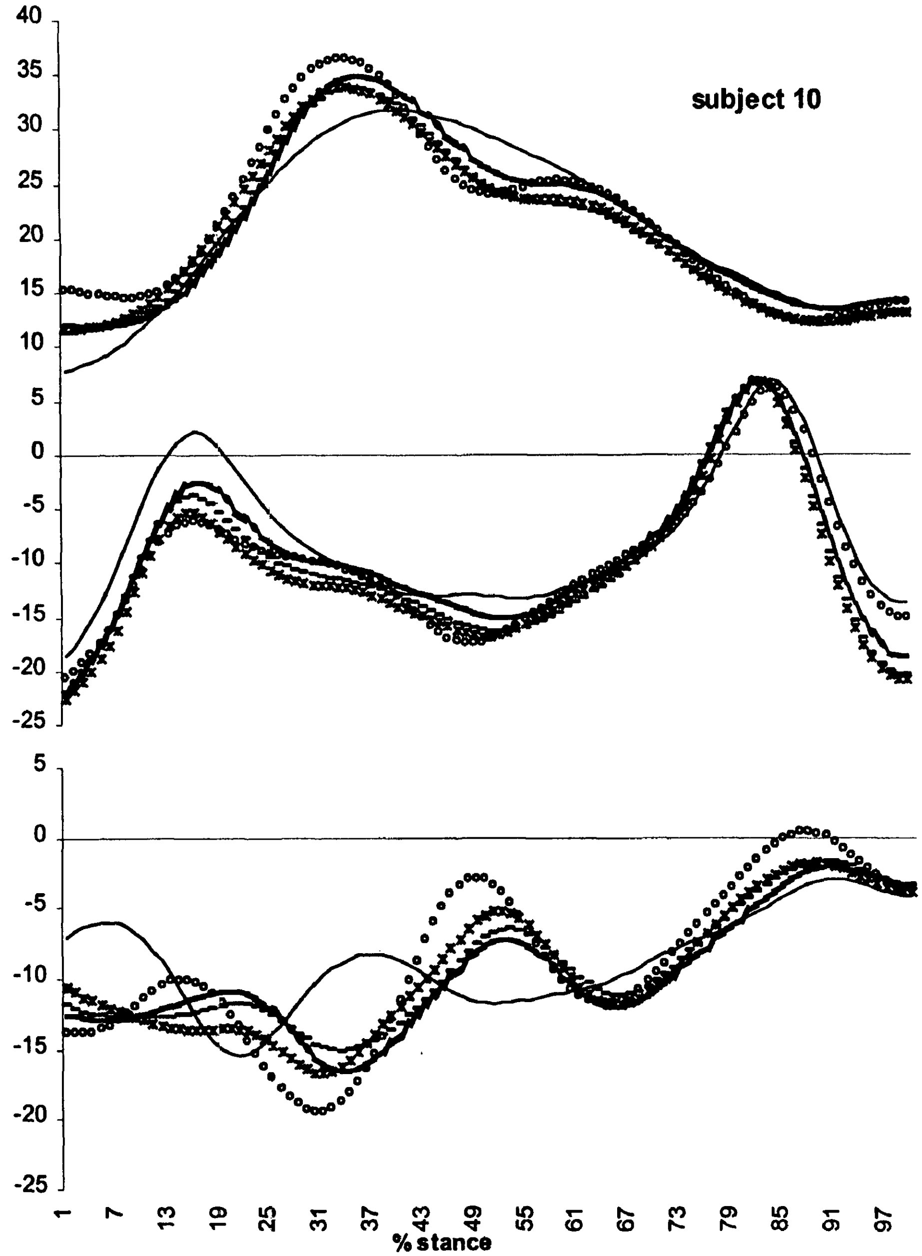
Figure E1. Pronation, knee, and tibial rotation mean ensemble curves for all conditions and subjects. The top group of curves are knee angles (+ = flexion), the middle group are pronation angles (+ = supination), and the bottom are tibial rotation angles (+ = internal rotation). *(Conditions appear as follows: no obstacle (0%) is the solid line, 5% is the coarse line, 7.5% is the dashed line, 10% is the asterisk line, and 12.5% is the circle line.)* (NOTE: Figure E1 continues to page 96.)





subject 5
subject 6
% stance


subject 8
subject 9
% stance
% stance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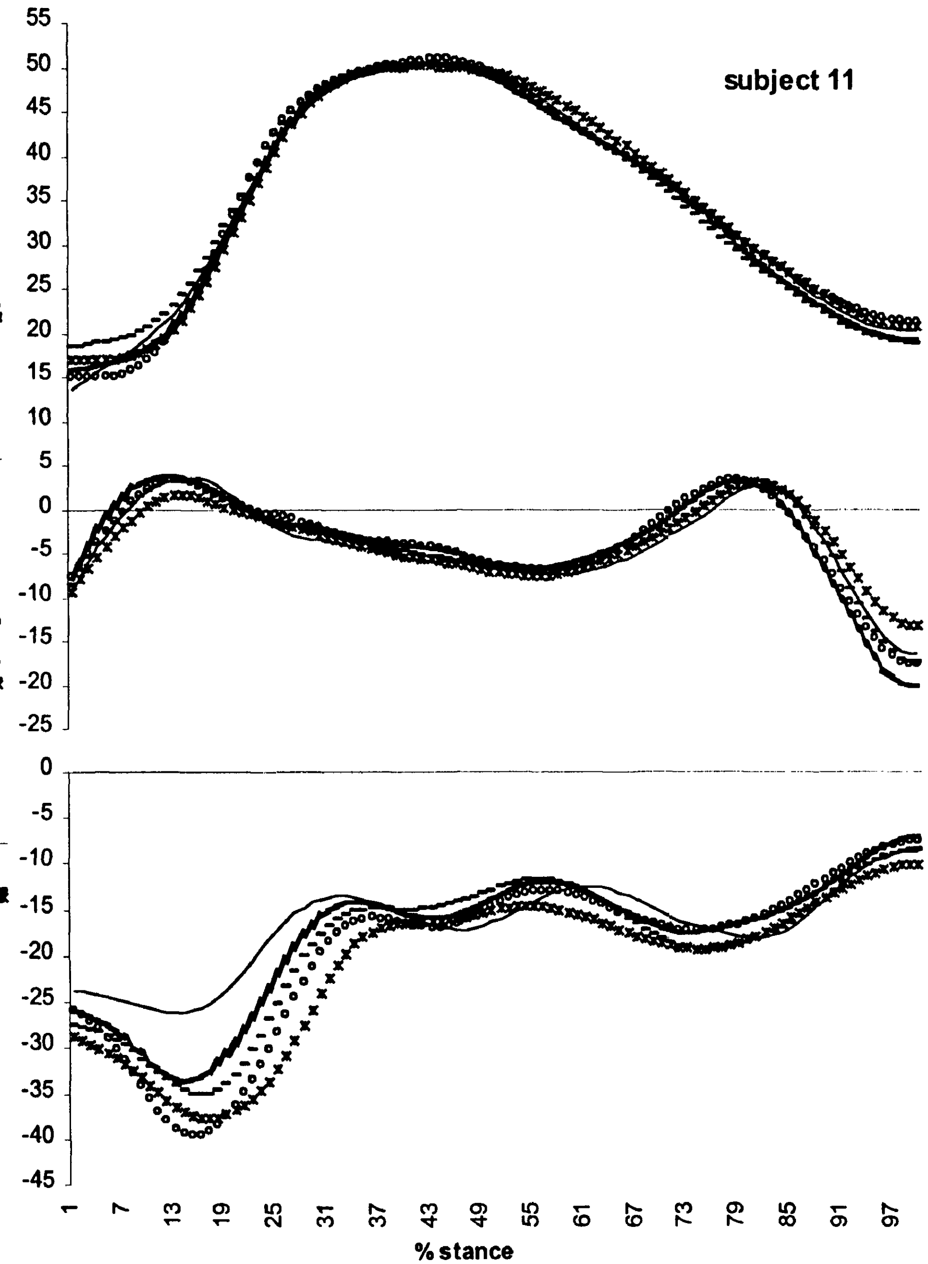
35
30
25
20
15
10
5
0
-5
-10
-15
-20
-25
5
0
-5
-10
-15
-20
-25
1 7 13 19 25 31 37 43 49 55 61 67 73 79 85 91 97
% stance


subject 11
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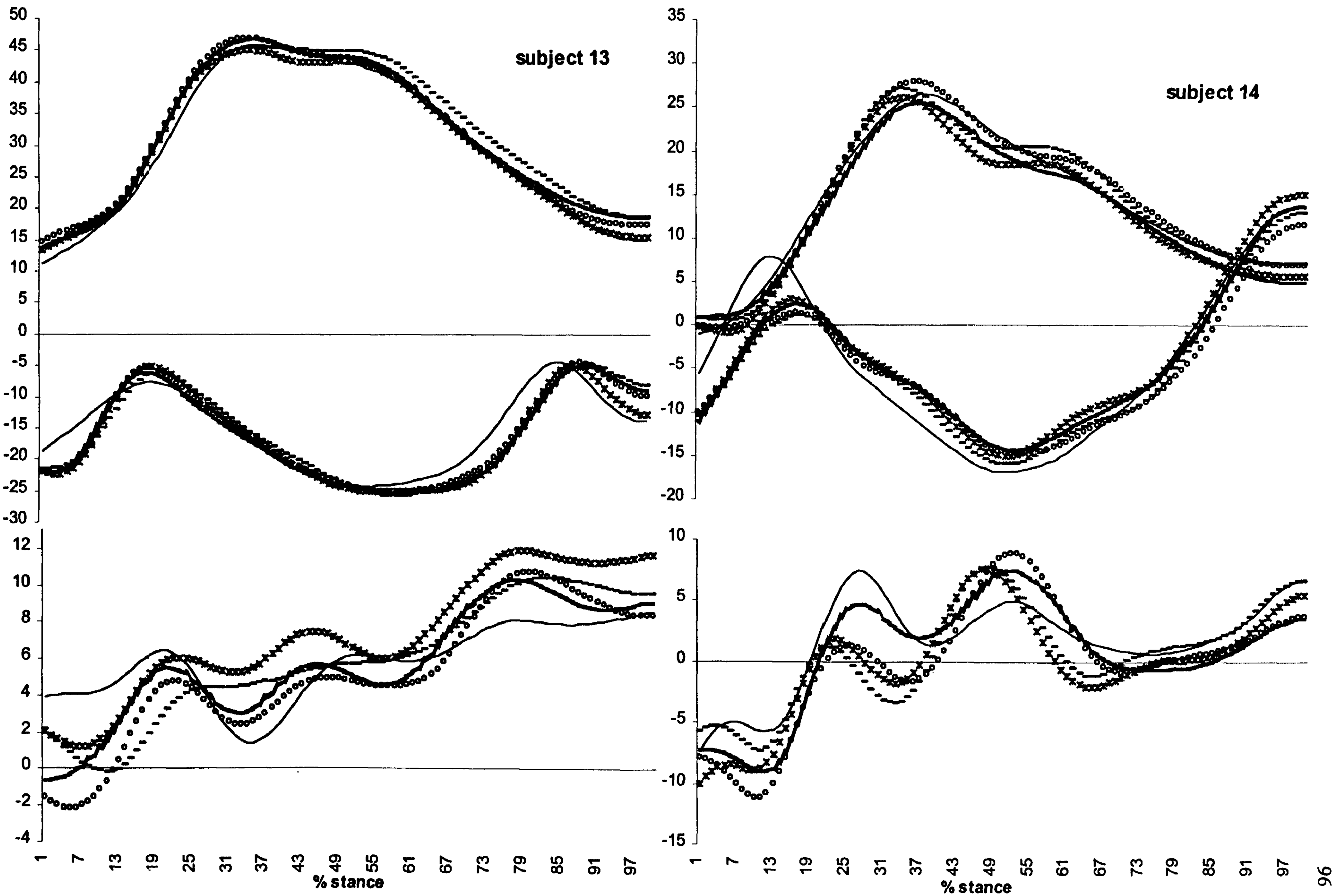
50
45
40
35
30
25
20
15
10
5
0
-5
-10
-15
-20
-25
0
-5
-10
-15
-20
-25
-30
-35
-40
-45
1 7 13 19 25 31 37 43 49 55 61 67 73 79 85 91 97
% stance


subject 13
subject 14
% stance
% stance