Factors influencing frontal plane knee joint kinematics and kinetics during running

ランニング中の前額面膝関節キネマティクス・キネティクスに影響を及ぼす因子

２０１３年１月

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Table of Contents

Chapter 1   Introduction

1-1. Preface

Running is one of the most common exercises for many individuals because of its

low cost, convenience and health benefits. However, the overall incidence rate of running

injuries has been estimated to be 29-56% per year (van Mechelen, 1992; Noguchi, 2000;

Taunton et al., 2002; Akimoto, 2007; van Gent et al., 2007). The most common site of running

injury is the knee, accounting for approximately 40% of running injuries, and patellofemoral

pain syndrome is the most prevalent injury among runners (Clement et al., 1981; Noguchi,

2000; Taunton et al., 2002; Akimoto, 2007). It is believed that the etiology of running injuries

is multifactorial, being related both to extrinsic and intrinsic factors (van Mechelen, 1992;

Noakes, 2002; Taunton et al., 2002).

A number of studies have linked running injuries to biomechanical variables, such as

running kinematics and kinetics (e.g. Stefanyshyn et al., 2006; Noehren et al., 2007). To date,

relationships between the anatomical structures and biomechanical variables remain uncertain.

In general, greater Q-angle is thought to be one of the contributing factors in the development

of running injuries at the knee. However, static measurements such as Q-angle are not always

representative of joint or segment kinematics during running (e.g. Heiderscheit et al., 2000).

Therefore, dynamic joint angles and joint loadings during running are deemed to be more important.

Both ankle and hip joint kinematics in the frontal and transverse planes are considered to influence frontal plane knee joint mechanics as the distal and proximal factors can lead to knee injuries during a closed kinetic chain exercise such as running, because, anatomically, there are limited adductor and abductor muscles around the knee. As a proximal factor, abnormal hip mechanics plays a crucial role in the development of knee injuries (Ireland et al., 2003; Powers, 2003; Dierks et al., 2008; Reiman et al., 2009). The weakness of the hip abductor and external rotator muscles may increase hip adduction, internal rotation, and subsequent knee abduction (Ireland et al., 2003; Powers, 2003). As well, with respect to a distal factor, excessive foot pronation (rearfoot eversion coupling with tibial internal rotation) has often been implicated in the development of running injuries at the knee (Tiberio, 1987; Hintermann and Nigg, 1998; Powers, 2003).

Even though both ankle and hip joint kinematics have been related with knee joint mechanics, it is unclear which of these factors are strongly related. Therefore, the purpose of this thesis was to elucidate possible mechanisms how ankle and hip joint kinematics associate with frontal plane knee joint mechanics during running.

1-2. Terminology

Frontal plane knee joint motion (knee abduction (valgus)/adduction (varus))

In previous studies, the terms "valgus/varus" and "abduction/adduction" have been used to represent frontal plane knee joint motion. Basically, both "valgus" and "abduction" refer to the outward angulation of the shank segment with respect to the thigh segment. "Valgus" often represents a position referred to as knocked-knee, and may occur from a pure "abduction" motion of the shank relative to the femur or from transverse plane knee rotation motions (Quatman and Hewett, 2009). Consequently, in this thesis, "knee abduction/adduction" is adopted as the representative term of frontal plane knee joint motion.

Rearfoot eversion

In previous studies, the terms "pronation/supination" and "eversion/inversion" have been used to represent frontal plane ankle joint motion. However, anatomically, foot pronation occurs around subtalar joint axis. The axis of subtalar joint runs upward, anteriorly in the sagittal plane and medially in the transverse plane from the long axis of the foot (Dugan and Bhat, 2005). Thus, foot pronation is defined as the tri-planar coupling movement of ankle (talocrural) joint dorsiflexion, calcaneus (subtalar) eversion and foot abduction. Due to the difficulty of measuring the movement of the talus with external skin markers, the entire ankle

joint complex has typically been simplified as a ball-and-socket joint (tibiocalcaneal joint)

with the Cardanic angles of plantar/dorsiflexion, abduction/adduction, and inversion/eversion

(e.g. Areblad et al., 1990; Reinschmidt et al., 1997). In this thesis, the term "rearfoot eversion"

is used to represent the movement of the rearfoot segment with respect to the shank segment in

the frontal plane.

Tibial internal rotation

In the calculation of three-dimensional ankle joint angle, transverse plane ankle joint

motion is calculated as the ankle external rotation or abduction, depending on the definition of

foot segment coordinate system. However, when the calcaneus (foot) is fixed to the ground

during the mid-stance of gait, it cannot abduct relative to the tibia (lower-leg). Thus, the tibia

(lower-leg) compensatory internally rotates (Tiberio, 1987; DeLeo et al., 2004). Consequently,

the term "tibial internal rotation" is adopted to describe inward rotation of the lower-leg with

respect to the foot segment.

1-3. Review of the relevant literature

This section will provide a brief review of running injury epidemiology and potential

etiological factors. A review of conceptual knee injury mechanisms will follow with specific

emphasis on distal (foot and ankle) and proximal (hip and pelvis) factors. Lastly, intra-limb coupling will be described.

1-3-1. Epidemiology and etiology of running injuries

Running provides many health benefits. Not surprisingly, there has been a dramatic increase in the number of recreational runners (Sasakawa Sports Foundation, 2010). However, it has been estimated that 29-56% of runners will sustain an injury during any 1-year period (the overall incidence rate of running injury was estimated from the following restricted studies which have samples of more than 300: Clement et al., 1981; van Mechelen, 1992; Noguchi, 2000; Taunton et al., 2002; Akimoto, 2007). The most common site of overuse running injuries is the knee, accounting for approximately 40% of all injuries, followed by foot and ankle, and lower leg (Clement et al., 1981; Noguchi, 2000; Taunton et al., 2002; Akimoto, 2007; van Gent et al., 2007). Notably, female runners are more likely to develop knee injuries (Taunton et al., 2002). The 5 most common overuse injuries are patellofemoral pain syndrome, accounting for approximately 25% of all injuries, followed to a much lesser degree by iliotibial band friction syndrome, plantar fasciitis, meniscal injuries and tibial stress syndrome (Taunton et al., 2002).

The high incidence of running injuries in the lower extremity has led to growing

interest in identifying the mechanical etiological factors associated with these injuries and their

mechanisms. It is believed that the etiology of running injuries is multifactorial, being related

both to extrinsic and intrinsic factors (van Mechelen, 1992; Noakes, 2002; Taunton et al.,

2002). The extrinsic factors include running mileage, intensity of exercise, running footwear

and running surface. The intrinsic factors are flexibility, anatomical structure, previous injury

and running experience. Nevertheless, a number of studies were conducted linking running

injuries with biomechanical variables (e.g. Stefanyshyn et al., 2006; Noehren et al, 2007). This

review will focus on the anatomical and biomechanical etiological factors and their

relationships with running injuries.

1-3-2. Anatomical measurements and running injury

Messier et al. (1988, 1991, 1995) and Duffey et al. (2000) evaluated several static

anatomical measures for particular running injuries including patellofemoral pain syndrome,

iliotibial band friction syndrome, shin splints, plantar fasciitis, Achilles tendonitis, and anterior

knee pain. In the study of Messier et al. (1991), the quadriceps femoris angle (Q-angle; Figure

1-1) was suggested as the only strong anatomical discriminator between runners afflicted with

patellofemoral pain syndrome and non-injured runners. In general, females have greater

Q-angle than males (Horton and Hall, 1989), and females are more likely to develop knee

injuries (Taunton et al., 2002). For these reasons, greater Q-angle is thought as one of the contributing factors in the higher incidence rate of running injuries at the knee in females. On the other hand, Duffey et al. (2000) reported that although subjects with anterior knee pain had high arches and smaller knee flexion, there was no significant difference in Q-angle between subjects with anterior knee pain and asymptomatic controls. Both prospective and retrospective studies have investigated the relationships between various static alignments including arch index, heel valgus, tubercle-sulcus angle, knee varus, and leg length difference and injury occurrence (Wen et al., 1997, 1998; Lun et al., 2006). Although a few significant relationships between static alignment measures with the risk of overuse running injury were found, the authors concluded that lower-extremity alignment is not a major risk factor for running injuries in their relatively low mileage cohort. In addition, Hreljac et al. (2000) compared several anatomical measures including longitudinal arch height, footprint index, and hamstrings and quadriceps femoris flexibility, and reported that there were no significant differences between a group of runners who had sustained at least one overuse running injury and a group of runners who had never sustained injury throughout their running careers in the selected anatomical measures. Willson and Davis (2008a) also reported that Q-angle and the ratio of pelvic width to femoral length between subjects with and without patellofemoral pain syndrome were similar. Based on the studies described above, it can be concluded that

anatomical measurements may play little role in the development of injury.

1-3-3. Biomechanical factors and running injury

Ankle joint (rearfoot) mechanics

In addition to investigating anatomical factors, biomechanical factors such as the kinematics and kinetics of the lower extremity during running have also been analyzed. One of the representative biomechanical risk factors of running injury is foot pronation. Pronation is defined as a combination of ankle dorsiflexion, subtalar eversion and forefoot abduction (Dugan and Bhat, 2005), and this movement is a necessary and protective mechanism during running because this mechanism allows impact forces to be attenuated over a long period (Ferber et al., 2009). The runners who demonstrate large pronation angles during running are classified as over-pronators (McClay and Manal, 1999; Ferber et al., 2009). Early studies reported that a high percentage of injured runners were over-pronators (James et al., 1978; Clement et al., 1981), so maximum pronation, pronation velocity, and time to maximum pronation have often been investigated as contributing factors to overuse running injuries. Excessive rearfoot frontal plane motion (eversion) is considered to influence lower extremity mechanics via tibial rotation (Tiberio, 1987; Powers, 2003).

Messier et al. (1991) compared subjects with patellofemoral pain syndrome with

asymptomatic controls for maximum pronation angle, velocity and time to maximum pronation. These authors, however, reported no significant differences between groups for any of the rearfoot kinematic measures. Duffey et al. (2000) also reported that there were no significant differences for these variables between subjects with anterior knee pain and healthy controls. In addition, Dierks et al. (2011) found that subjects who have patellofemoral pain exhibited smaller peak rearfoot eversion compared to controls. Similarly, the retrospective and prospective studies on iliotibial band friction syndrome found no difference in rearfoot eversion between injured and non-injured runners (Noehren et al., 2007; Ferber et al., 2010). These studies suggest that the influence of foot pronation and/or rearfoot eversion on knee joint injuries is questionable.

In contrast, Willems et al. (2007) prospectively evaluated exercise-related lower leg pain in 400 young subjects for 1 academic year. They collected foot plantar pressure and three-dimensional rearfoot kinematics during running. Seventy-five runners developed lower leg pain, and injured runners exhibited significantly greater pronation excursion and increased planter pressure underneath medial side of foot. Excessive rearfoot movement has also been suggested as a risk factor for plantar fasciitis and Achilles tendinitis (Kosmahl and Kosmahl, 1987; Kvist, 1994). These results suggest that rearfoot motion may be related to the possibility of developing injuries at the foot and ankle.

Knee joint mechanics

In general, greater knee valgus (abduction) is considered to be a risk factor of patellofemoral pain syndrome (Powers, 2003). This motion may increase the dynamic Q-angle. An in-vitro cadaveric study reported that a greater dynamic Q-angle could lead to lateral patellar dislocation and increased lateral patellofemoral contact pressures (Mizuno et al., 2001). Therefore, it has been considered that the greater Q-angle in females could lead to greater dynamic Q-angle and/or knee valgus (abduction), and a high incidence rate of knee injury in females (Ferber et al., 2003).

Despite the high incidence of running injuries at the knee, only one study prospectively and retrospectively investigated knee joint loading for patellofemoral pain syndrome (Stefanyshyn et al., 2006). This study found that subjects with patellofemoral pain syndrome demonstrated increased internal knee abduction angular impulse compared to non-injured controls during running. Increased knee abduction moments during running could be generated by increased muscle forces and/or increased soft tissue forces. Increased knee joint loading magnitudes, as indicated by increased joint impulses, would likely to reflect increase in loads and stresses on the lateral facet of the patella, and result in injuries after repetitive cycles.

Hip joint mechanics

Recently, hip joint kinematics and the muscles that control and stabilize the pelvis have received attention in association with the development of running injuries at the knee. Poor pelvic muscular control has been implicated to contribute to patellofemoral malalignment and subsequent patellofemoral pain (Fulkerson, 2002; Powers, 2003). Altered patellofemoral tracking may be the result of the femur rotating medially beneath the patella rather than the patella moving laterally on the femur during weight-bearing activities (Powers, 2003). The hip muscles control pelvic stability and leg alignment, so the absence of sufficient hip muscle strength, particularly the hip abductors and external rotators, may lead to excessive femoral internal rotation and hip adduction movements and subsequent knee valgus motion during weight-bearing activities (Ireland et al., 2003; Powers, 2003).

In fact, several studies commonly reported that subjects with patellofemoral pain syndrome or iliotibial band friction syndrome demonstrated greater hip adduction than asymptomatic control subjects (Noehren et al., 2007; Willson and Davis, 2008a; Ferber et al., 2010). However, several conflicting studies exist regarding greater hip internal rotation. Souza and Powers (2009) reported that female subjects with patellofemoral pain demonstrated greater hip internal rotation while performing a variety of tasks including running. In contrast, Willson and Davis (2008a) reported greater hip adduction but not greater hip internal rotation while

running. On the other hand, Dierks et al. (2008) reported that subjects with patellofemoral pain had less hip adduction and no differences in hip internal rotation when compared to non-injured controls. Although hip internal rotation has been considered to affect knee valgus (abduction) motion, there is no consensus about hip internal rotation on knee joint injuries.

The relationships between anatomical measures and biomechanical factors

Several studies have focused on the gender differences in knee and hip mechanics during running (Malinzak et al., 2001; Ferber et al., 2003; Schache et al., 2003; Chumanov et al., 2008). A common observation among these studies is that female runners demonstrate greater hip adduction and internal rotation, as well as greater knee abduction throughout the stance phase (Malinzak et al., 2001; Ferber et al., 2003; Chumanov et al., 2008). Lower extremity kinematic profiles of female runners are considered to be one of the factors explaining the gender-related discrepancy in the incidence rate of knee injuries. In general, females have anatomical structural differences compared to males (Horton and Hall, 1989), and these anatomical structural differences are thought to affect the gender differences in running kinematics.

However, Willson and Davis (2008a) reported that Q-angle and the ratio of pelvic width to femoral length between subjects with patellofemoral pain syndrome and controls

were similar, but the injured runners demonstrated significantly greater hip adduction and smaller hip internal rotation than uninjured controls. On the other hand, Heiderscheit et al. (2000) investigated the influence of Q-angle on the three-dimensional lower-extremity kinematics during running and did not reveal any differences between high and low Q-angle groups in maximum joint or segment angles. Static measurements such as Q-angle are not always representative of joint or segment kinematics during running.

1-3-4. Potential knee injury mechanisms

Conceptually, both ankle and hip joint kinematics are considered to influence knee joint kinematics and can lead to knee injuries during a closed kinetic chain exercise such as running (Tiberio, 1987; Powers, 2003; Noehren et al., 2007; Dierks et al., 2008). However, it is not clear yet whether both proximal and distal factors synergistically affect knee joint kinematics or not.

Distal factor

With respect to distal factors, excessive foot pronation has often been implicated with patellofemoral pain because the patella tendon attaches to the tibia and pronation is coupled with tibial rotation. Tiberio (1987) proposed a mechanism of foot pronation and its

influence on patellofemoral mechanics. To achieve knee extension during mid-stance of gait, the tibia must externally rotate with respect to the femur to ensure adequate motion for the screw-home mechanism (Powers, 2003). However, if an individual has excessive pronation or exhibits pronation beyond mid-stance, the tibial external rotation that is coupled with foot supination might be delayed, and subsequently, compensatory internal rotation of the femur would be required in order to minimize trauma at the tibiofemoral joint. The tibiofemoral joint may be protected from injury by this compensatory mechanism, but it is not beneficial for the patellofemoral joint. When the femur internally rotates, the compression between the lateral facet of the patella and the lateral femoral condyle would be increased. Therefore, it is believed that foot pronation may play a role in the development of patellofemoral dysfunction.

Proximal factor

As a proximal factor, abnormal hip mechanics may play a crucial role in the development of knee injuries (Powers, 2003; Dierks et al., 2008). Weakness of the hip abductor and external rotator muscles may increase hip adduction, internal rotation, and subsequent knee abduction (Ireland et al., 2003; Powers, 2003). Excessive femoral adduction and internal rotation during dynamic tasks can be the result of weakness of the hip abductors and external rotators (Powers, 2003). If these muscles are weak, this may contribute to

excessive thigh adduction and internal rotation, and subsequent knee valgus (abduction) during

weight-bearing activities.

1-3-5. Intra-limb coupling

The stance phase of running may be divided into two functional phases (DeLeo et al.,

2004). The first half of stance is commonly referred to as the cushioning, or eccentric phase of

gait. The last half of stance is referred to as the propulsion, or concentric phase. Generally,

following heel strike, the knee flexes, the tibia internally rotates and the foot pronates. When

the calcaneus is fixed to the ground, it cannot abduct relative to the talus (Tiberio, 1987;

DeLeo et al., 2004). Therefore, the talus compensatory adducts. Due to the tight articulation of

the ankle joint mortise, the tibia internally rotates as the talus adducts. Inversely, following

mid-stance, the knee should extend, the tibia externally rotates and the foot supinates. During

the first half of stance, the knee joint flexes which is also associated with tibial internal rotation.

Hence, pronation, tibial internal rotation, and knee flexion occur relatively synchronously

(DeLeo et al., 2004).

Tiberio (1987) proposed that if runners demonstrate excessive pronation or

prolonged pronation during the stance phase of running, the external rotation of the tibia that is

coupled with foot supination would be delayed, and subsequently the femur would be

internally rotated in order to minimize trauma to the tibiofemoral joint and to achieve knee extension. However, this protective mechanism for the tibiofemoral joint may increase the patellofemoral joint contact pressure. Thus, the decoupling of lower extremity kinematics is considered as one potential cause for running injury. The joint coupling variables were mainly determined as the ratio of peak rearfoot eversion to peak tibial internal rotation (e.g. McClay and Manal, 1997; Stacoff et al., 2000). Although both ankle and hip joint kinematics are considered to influence knee joint kinematics, the couplings between knee frontal plane motion, which is the kinematic risk factor at the knee, and the other joint motions have not been addressed.

1-4. Purpose

Based on the studies described in the review of the literature, the relationships between the anatomical structures and biomechanical variables remain uncertain. Hence, dynamic joint and segment angles and joint loadings during running are deemed to be more important for understanding running injury mechanisms. Recently, both ankle and hip joint kinematics are considered to influence knee joint mechanics during running. Although this has been proposed as a "concept" for knee injury mechanism in several studies, it has not been well examined so far. Therefore, it is not clear yet whether both foot/ankle and hip joint

kinematics associate with knee joint mechanics or not. The general purpose of this thesis is to elucidate possible mechanisms how ankle and hip joint kinematics associate with knee joint mechanics during running. The outlines of each chapter are as follows.

In Chapter 2, gender differences in lower extremity kinematics during running were examined, with specific emphasis on the relationships between ankle and hip joint kinematics and knee joint kinematics in the frontal plane.

In Chapter 3, the associations of ankle and hip joint kinematics with kinematic risk factors of running injury at the knee were investigated to elucidate how distal and proximal factors are related with frontal plane knee joint motions.

In Chapter 4, the relationships between lower extremity kinematics and frontal plane knee joint loading were investigated focusing on the relationships between toe-out and hip rotation angles and frontal plane lever arm at the knee.

In Chapter 5, the main findings of each chapter were summarized. Then, the influences of ankle and hip joint kinematics on both kinematic and kinetic risk factors of running injury at

the knee were discussed to identify and explain the mechanisms. Lastly, several factors that

might affect the interpretation of the present results were addressed.
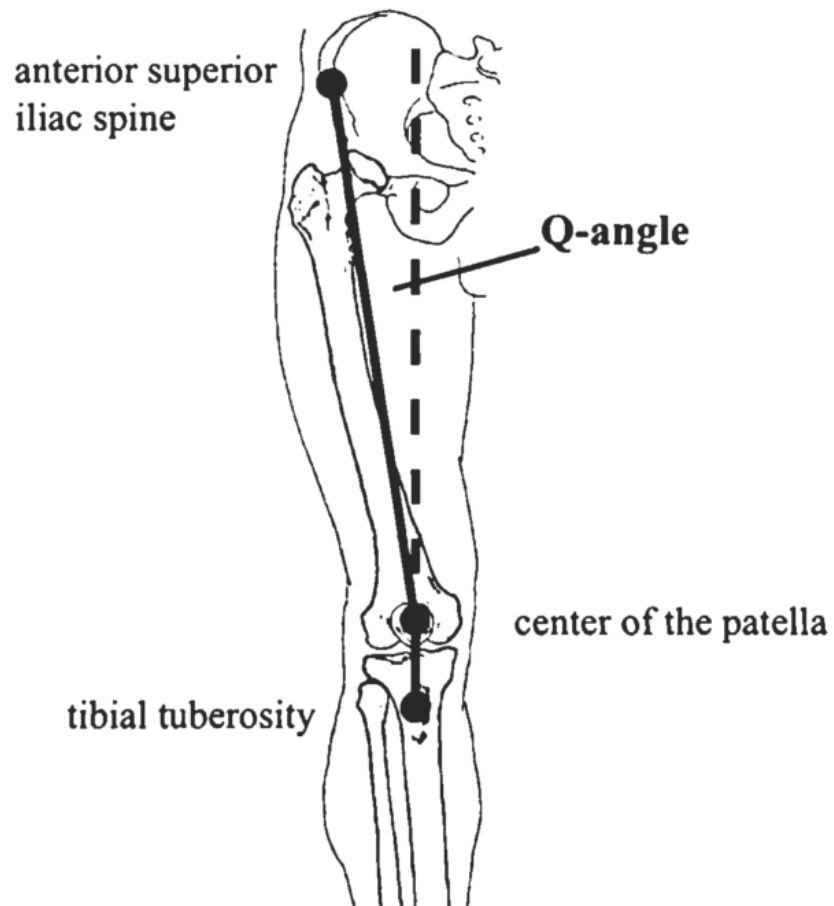
Figure 1-1. Quadriceps femoris angle (Q-angle). Q-angle is defined as the angle between a line connecting the anterior superior iliac spine and the center of the patella and a line connecting the center of the patella and the tibial tuberosity (From Heiderscheit et al., 2000).
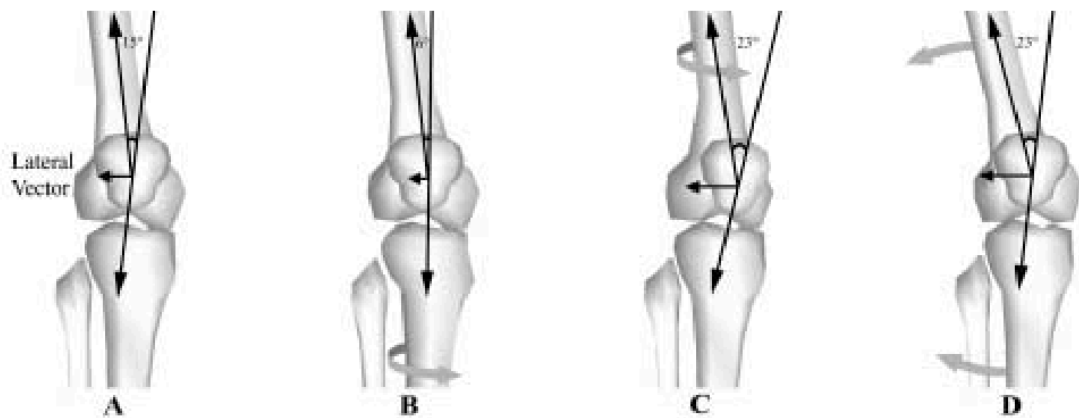
Figure 1-2. Q-angle and resultant of the forces applied to the patella.
(A) Normal alignment of the tibia and femur results in an offset in the resultant quadriceps force vector (proximal) and the patellar tendon force vector (distal), creating a lateral vector acting on the patella.
(B) Tibia internal rotation decreases the Q-angle and the magnitude of the lateral vector acting on the patella.
(C) Femoral internal rotation increases the Q-angle and the lateral force acting on the patella.
(D) Knee valgus increases the dynamic Q-angle and the lateral force acting on the patella.
(From Powers, 2003)
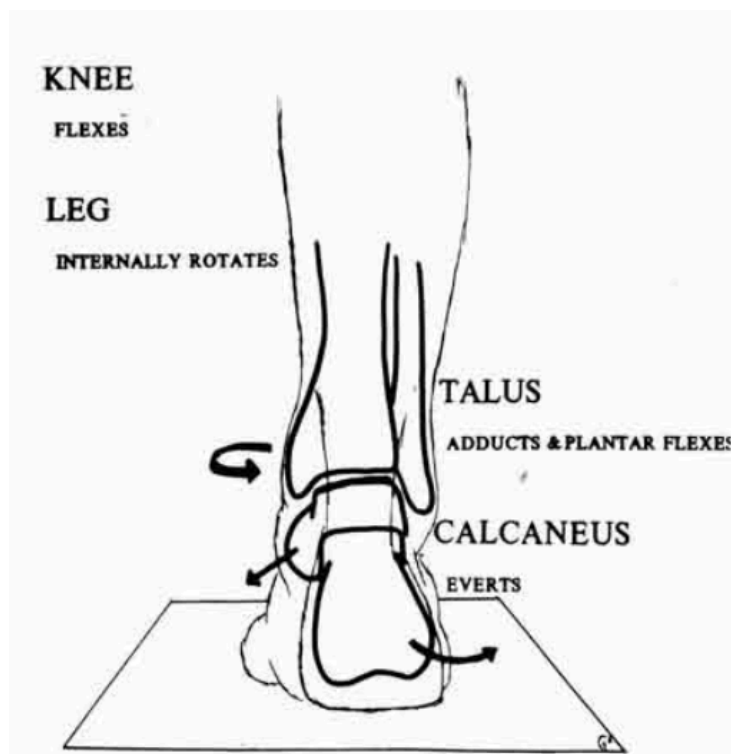
Figure 1-3. Internal rotation of the lower leg with closed chain foot pronation (right leg).
When the subtalar pronates during ground contact, the calcaneus everts and the head of the talus slides medially, and subsequently the lower-leg internally rotates (From Tiberio, 1987).
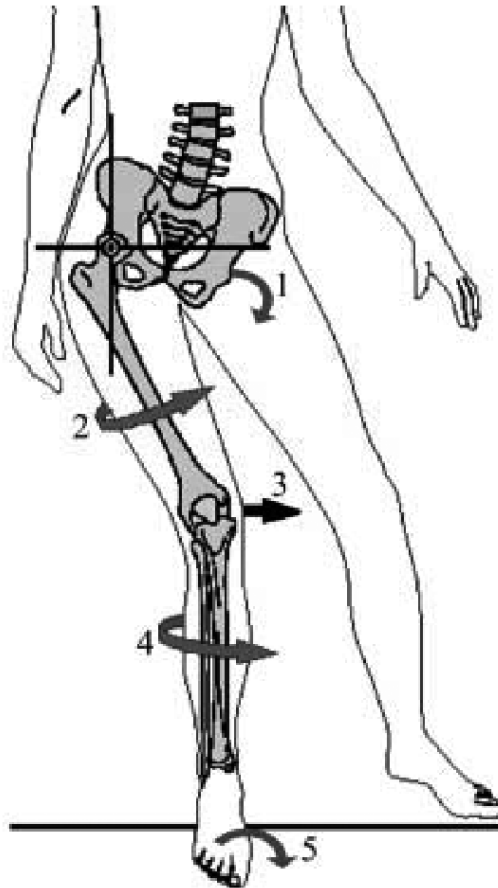
Figure 1-4. Schematic showing the potential contributions of the various lower-extremity segments to abnormal alignment.
(1) contralateral pelvic drop, (2) femoral internal rotation, (3) knee valgus, (4) tibia internal rotation, and (5) foot pronation
(From Powers, 2003)

Figure 1-5. Lower extremity joint angles during stance phase of running.
(a) rearfoot eversion/inversion, (b) tibial rotation,
(c) knee flexion/extension, and (d) hip rotation.
Lines indicate the peak timing of each variable (From DeLeo et al., 2004)

Chapter 2   **Gender differences in hip and ankle joint kinematics on knee abduction during running**

2-1. Introduction

The most common site of running injuries is the knee, followed by foot and ankle, and lower leg (Taunton et al., 2002). Notably, female runners are mo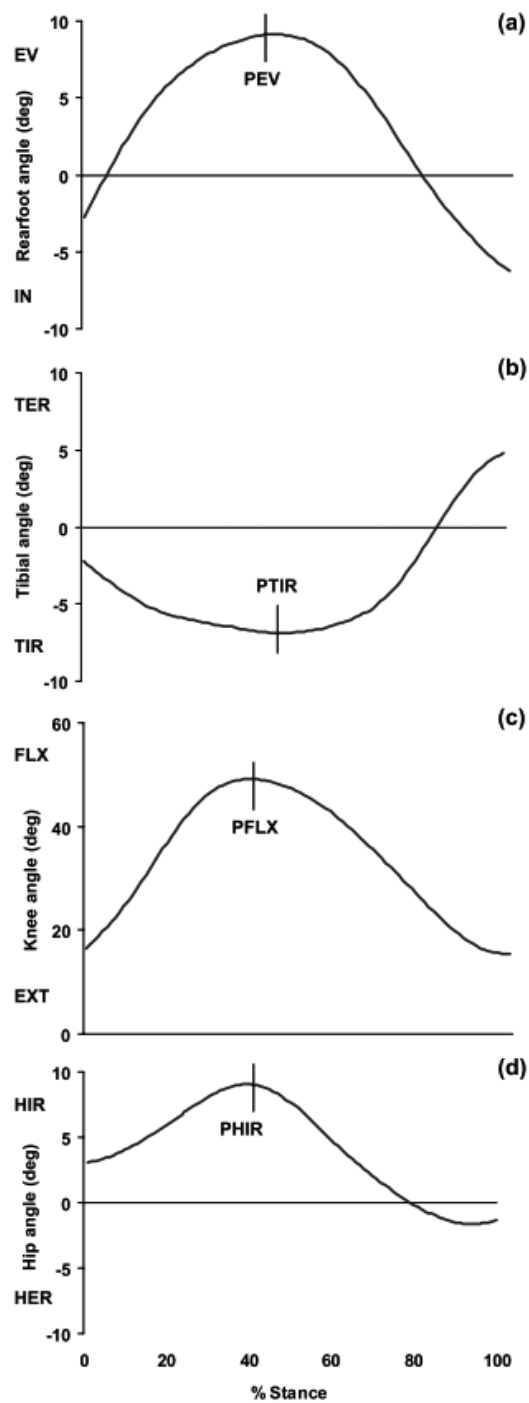re likely to develop knee injuries, such as patellofemoral pain syndrome and iliotibial friction syndrome (Taunton et al., 2002). Several studies have focused on the gender differences in knee and hip mechanics during running (Malinzak et al., 2001; Ferber et al., 2003; Schache et al., 2003; Chumanov et al., 2008). A common observation among these studies is that female runners demonstrate greater hip adduction and internal rotation, as well as greater knee abduction throughout the stance phase (Malinzak et al., 2001; Ferber et al., 2003; Chumanov et al., 2008). These profiles of female runners in the lower extremity kinematics are considered to be one of the factors explaining the gender-related discrepancy in the incidence rate of knee injuries.

Both ankle and hip joint kinematics are considered to influence knee joint kinematics and can lead to knee injuries during a closed kinetic chain exercise such as running (Noehren et al., 2007; Dierks et al., 2008). However, it is not clear yet whether both proximal and distal factors are associated with knee abduction or not. In addition, no study has ever

attempted to simultaneously compare the ankle, knee and hip joint kinematics during running between females and males. It is possible that gender differences in lower extremity kinematics, at the ankle in particular, may also be a factor related with the gender discrepancy of incidence rate of running injuries. Therefore, the purpose of this study was to clarify gender differences in the lower extremity kinematics during running, with a specific emphasis on the relationships between the distal and proximal factors and knee joint kinematics. Based on the previous studies on proximal and distal factors on knee abduction, it is hypothesized that female runners would exhibit greater peak knee abduction associated with greater rearfoot eversion and greater hip adduction than male runners.

2-2. Methods

Subjects

A priori statistical power analysis ($\alpha = 0.05$, 80% power) using maximal knee abduction angle obtained in a pilot study (6 females, 6 males) indicated that a minimum of 22 subjects (11 females, 11 males) were required for this investigation. Based on this, 11 female (mean age: 20.7 ± 0.8 years, height: 1.60 ± 0.05 m, mass: 52.1 ± 6.1 kg, weekly running distance: 36.5 ± 30.1 km; mean $\pm$ SD) and 11 male (mean age: 22.0 ± 1.7 years, height: 1.73 ± 0.07 m, mass: 62.8 ± 6.5 kg, weekly running distance: 42.0 ± 41.4 km; mean $\pm$ SD)

recreational runners were recruited to participate in this study. All subjects had no lower

extremity injury within six months prior to the measurement, kept habitual running experience

of at least 10 km/week, and did not participate in any organized training system of competitive

long distance runners. This study was approved by Human Research Ethics Committee in the

Faculty of Sport Sciences, Waseda University. Informed written consent was obtained from all

subjects.

Data Collection

Three-dimensional kinematics of the right leg were analyzed. During data collection,

all subjects wore the identical-model running shoes (Adizero Boston, adidas AG, Germany)

with a moderate cushioning property. Twelve spherical retro-reflective markers were attached

on the skin of the pelvis, thigh, shank and rearfoot. Three markers per segment were used to

track the position and orientation of the segments, attached to the following locations: sacrum

which is the midpoint of posterior superior iliac spine, right anterior superior iliac spine

(ASIS), left ASIS, proximal lateral thigh below the greater trochanter, distal lateral thigh, distal

anterior thigh, proximal lateral shank, midtibial crest, distal lateral shank, upper shoe heel,

lower shoe heel, and lateral side of the shoe below the lateral malleoli (Figure 2-1). When

attaching the tracking markers, the subjects were requested to contract the muscles of the right

leg several times in the standing position in an attempt to avoid the areas of large muscle deformation.

Prior to the running trial, a standing neutral trial was captured (1 second). In this trial, the subjects were requested to stand in a position with the feet standardized to the laboratory coordinate system and approximately hip width apart. The knee and hip were in a fully extended position, with the ankle joint at approximately a 90° angle. For the neutral trial, additional anatomical markers were placed on the following locations to define the joint centers and anatomical segment coordinate systems: greater trochanter, lateral femoral epicondyle, patella, medial and lateral malleoli, the heads of the 1st and 5th metatarsals and the tip of the shoe. After the neutral trial was finished, the anatomical markers were removed.

In the running trials, subjects ran along a 25m runway at a speed of 3.5 m/s. Subjects were given several practice trials to ensure that the foot landed with a natural running style on the force plate. The running speed was monitored with photocells (E3G-MR19T, Omron Corp., Tokyo, Japan) placed just before and after the force plate at a distance of 1.92 m. Ten successful trials were collected during the stance phase of running. Those trials in which (a) the running speed was not within 5% of the target running speed and/or (b) the subject could not contact the force plate correctly with a natural running style were rejected. A motion capture system with eight infrared cameras (Motion Analysis Corp., Santa Rosa, CA) was used

to determine three-dimensional marker positions at 240 Hz. Ground reaction forces were simultaneously collected at 2400 Hz using a force plate (Bertec Corp., Columbus, OH) to determine heel-strike and toe-off for the identification of the stance phase. Before testing, cameras were calibrated to a defined capture volume (2.5 m length × 1.2 m width × 1.5 m height), and the average three-dimensional residual was below 0.5 mm.

Data Reduction

All kinematic data were analyzed for the stance phase. Custom-made software developed on the Matlab platform (R2008b, Mathworks Inc., Natick, MA) was used for the processing and analysis. Before calculation, the kinematic data were filtered using a fourth-order low-pass Butterworth filter with a cut-off frequency of 12 Hz (Pohl et al., 2006; Stefanyshyn et al., 2006). A threshold of vertical ground reaction force was set at 20 N to determine heel strike and toe-off. The marker position data of running trials were tracked for a period corresponding to 20 frames before and after contact with the force plate.

A right-handed orthogonal coordinate system embedded to each segment was defined by anatomical markers and joint centers. Initially, the pelvis coordinate system was defined by the right and left ASISs and sacrum markers (Wu et al., 2002). After the pelvis coordinate system was defined, the hip joint center was estimated based on the inter-ASIS

distance (Bell et al., 1989). The antero-posterior coordinate of the hip joint center was defined by the greater trochanter marker (Stefanyshyn et al., 2006). The knee joint center was identified by markers on the lateral femoral epicondyle and the patella (Stefanyshyn et al., 2006). The midpoint of two markers on the lateral and medial malleoli was used to identify the ankle joint center. Once joint centers were defined, the thigh and shank coordinate systems were defined by the proximal and distal joint centers and lateral marker of the proximal joint center of each segment (greater trochanter for the thigh, lateral femoral epicondyle for the shank). The foot coordinate system was considered aligned with the laboratory coordinate system. During running trials, the anatomical markers were removed because these markers were sliding on the bony landmarks due to the skin deformation, and the tracking markers were used to track the position and orientation of the segments. In this calculation, the singular value decomposition method (Söderkvist and Wedin, 1993) was utilized to optimally determine the anatomical segment coordinate systems for the thigh, shank and foot segments during running trials (Chiari et al., 2005).

Joint angles were determined as the orientation of the distal segment with respect to the proximal segment according to the definition of the joint coordinate system (Grood and Suntay, 1983; Cole et al., 1993). In addition, step width was calculated from the marker placed on the sacrum and ankle joint center at an instant when the peak vertical ground reaction force

was attained. Both sacrum marker and ankle joint center were projected onto the frontal plane

of the laboratory coordinate system, and then the perpendicular distance, which was defined as

the step width in this study, was measured from the ankle joint center to a vertical line drawn

from the sacrum marker. The step width determined for each trial was normalized by the

subject's ASIS width.

The following variables were extracted from the time-series data, the peak values of

rearfoot eversion, tibial internal rotation, knee abduction, knee internal rotation, hip adduction

and hip internal rotation. To compare the time course changes in joint angles between trials

and among subjects, the time-series joint angle data were time normalized to 101 data points,

with each time interval representing 1% of the stance phase. The instant at which peak joint

angle was recorded for each joint was also compared between genders.

Statistical Analyses

Descriptive data are presented as mean and standard deviation. Statistical analyses

between male and female runners were performed using independent t-tests. To quantify the

magnitude of differences between genders, the effect size (ES) was calculated for each

parameter using Cohen's d. The ES values of less than 0.5 represent small differences, 0.5 to

0.8 represent medium differences, and greater than 0.8 represent large differences (Cohen,

1988). The correlation between peak rearfoot eversion and peak knee abduction and peak hip adduction and peak knee abduction were determined by Pearson's correlation coefficient. Additionally, the interclass correlation coefficients (ICC) were calculated to determine the trial-retrial repeatability for the variables of interest. To establish the testing reliability, five male subjects were chosen to repeat the testing procedure within 1 month after the data collection. For all analyses, the significance level was set at $\alpha \leq 0.05$. Statistical analyses were performed using SPSS 12.0 J (SPSS Inc., Chicago, IL). For all analyses, the significance level was set at $\alpha \leq 0.05$. Statistical analyses were performed using SPSS 12.0 J (SPSS Inc., Chicago, IL).

2-3. Results

There was no significant difference between female (227 ± 22 ms) and male runners (228 ± 22 ms) in the stance duration ($P > 0.05$). Figure 2-2 shows the ensemble average of three-dimensional angular motions of the ankle, knee and hip joints for female and male runners during stance phase of running. Throughout the stance phase, female runners demonstrated similar flexion/extension angles at the ankle, knee and hip joint compared to those of male runners. In both female and male runners, the rearfoot landed in inversion and then immediately moved into eversion. Hip adduction increased immediately after heel-strike,

and subsequent peak knee abduction occurred at approximately mid-stance. The hip internal rotation occurred around heel-strike, and then hip joint rotated externally during weight-bearing.

Mean values of the peak ankle, knee and hip joint angles, instants at which the peak joint angles were recorded for each joint and normalized step width are listed in Table 2-1. At the hip, female runners demonstrated greater hip adduction ($P < 0.01$) and internal rotation ($P < 0.05$) than male runners. At the knee, female runners demonstrated greater knee abduction ($P < 0.05$), whereas male runners demonstrated a trend toward greater knee internal rotation ($P = 0.094$). At the ankle, male runners demonstrated a greater rearfoot eversion ($P < 0.01$). Tibial internal rotation and normalized step width had no effect of gender ($P > 0.05$). There were no significant differences between genders in the instant at which the peak joint angle was recorded for each joint ($P > 0.05$).

The relationships between the knee joint kinematics and rearfoot eversion and hip abduction are presented in Figure 2-3. In all 22 subjects, the peak knee abduction was positively correlated with the peak hip adduction ($r = 0.495$, $P < 0.05$; Figure 2-3 (A)), and negatively correlated with the peak rearfoot eversion ($r = -0.685$, $P < 0.001$; Figure 2-3 (B)).

The trial-retrial ICC values ranged from 0.863 to 0.989, and the test-retest ICC values ranged from 0.776 to 0.983 for all kinematic variables of interest. The mean value for

the root mean square (RMS) differences calculated between trial and retrial data was less than 1.6°, and the mean value for the RMS differences between test-retest data was less than 1.7°.

2-4. Discussion

This study found that female runners demonstrated greater peak knee abduction, hip adduction and hip internal rotation during the stance phase of running, whereas male runners demonstrated greater peak rearfoot eversion. In addition, the peak knee abduction was correlated positively with peak hip adduction and negatively with peak rearfoot eversion. The results of this study did not support our hypothesis that female runners would exhibit greater knee abduction associated with greater rearfoot eversion and greater hip adduction than male runners. As shown in Figure 2-3, the most extreme cases for female and male runners were plotted at similar positions in the graphs.

Contrary to our hypothesis, rearfoot eversion was found to be 5° greater in the male runners during running. The rearfoot eversion, which results in increased tibial internal rotation, is considered to lead to the knee abduction (Powers, 2003). In this study, the rearfoot landed in inversion and then immediately moved into eversion within 20-30% of stance phase, and then peak rearfoot eversion and peak knee abduction occurred at approximately 45% of stance phase in both female and male runners. Although large rearfoot eversion movement

occurred prior to knee abduction, the larger peak knee abduction did not correspond to larger peak rearfoot eversion. However, greater rearfoot movement may be a risk factor of running injuries at the foot, ankle and lower leg, such as tibial stress fracture (Pohl et al., 2008), Achilles tendonitis (Kvist, 1994) and plantar fasciitis (Kosmahl and Kosmahl, 1987). The greater rearfoot eversion observed in male runners is in line with a fact that male runners are more prone to Achilles tendonitis and plantar fasciitis than females (Taunton et al., 2002).

Female runners demonstrated significantly greater peak hip adduction and hip internal rotation than male runners during the stance phase of running. These findings are consistent with previous studies (Malinzak et al., 2001; Ferber et al., 2003; Chumanov et al., 2008). Simultaneous increases in the hip adduction and internal rotation have been suggested as contributing factors in running injuries at the knee (Ireland et al., 2003; Powers, 2003; Cichanowski et al., 2007) and other common running injuries (Niemuth et al., 2005; Pohl et al., 2008). In fact, several studies commonly reported that the subjects with patellofemoral pain syndrome or iliotibial band friction syndrome demonstrated greater hip adduction than asymptomatic control subjects (Noehren et al., 2007; Willson and Davis, 2008a, 2008b; Ferber et al., 2010). Weaker hip strength has been suggested to contribute to an increase in hip adduction and internal rotation (Ireland et al., 2003; Leetun et al., 2004; Niemuth et al., 2005; Cichanowski et al., 2007). Leetun et al. (2004) prospectively compared the differences in hip

34

strength between male and female athletes as well as between those athletes who were later injured and those who were not. In their results, male athletes had stronger hip abductor and external rotator strength compared with females. In addition, the injured athletes showed significantly weaker hip abductor and external rotator strength than uninjured athletes. Collectively, weaker hip strength may be related to greater hip adduction and internal rotation, particularly in females. Although the present study cannot give a direct link between joint kinematics and the incidence of injury, previous findings support the notion that the hip joint kinematics are partially associated with the higher incidence rate of knee injuries in female runners.

In our study, female runners demonstrated greater peak hip internal rotation than male runners, but the peak hip internal rotation occurred at heel-strike, and then the hip joint rotated externally. Although the profile of hip motion in the transverse plane differed from previous studies (Ferber et al., 2003; Chumanov et al., 2008), females demonstrated a tendency towards a more internally rotated hip position than males. This fact is consistent with previous studies (Ferber et al., 2003; Chumanov et al., 2008). However, inconsistent observations were reported regarding hip internal rotation between the subjects with patellofemoral pain syndrome and uninjured controls (Dierks et al., 2008; Willson and Davis, 2008a; Souza and Powers, 2009). Also, there were no significant differences in tibial internal

rotation and knee internal rotation between genders in the present study. The findings of our study suggest that the frontal plane joint motions such as hip adduction, knee abduction and rearfoot eversion have greater influence to explain the gender discrepancy of the incidence rate of running injuries than the transverse plane joint motions such as internal/external rotation at the ankle, knee and hip joints.

The main finding of this study was that there was a significant positive correlation between the peak knee abduction and the peak hip adduction, while there was a significant negative correlation between the peak knee abduction and the peak rearfoot eversion. In other words, both proximal and distal factors do not synergistically increase knee abduction. Conceptually, the rearfoot eversion during weight-bearing induces tibial internal rotation, and subsequently knee abduction (Powers, 2003). Similarly, hip adduction also has been suggested to induce knee abduction. However, the result of this study contradicted this concept on the influence of rearfoot eversion. Based on the current results, we may say that the relationships among the rearfoot eversion, knee abduction and hip adduction for a given foot position relative to the pelvis are compensatory joint motions. A schematic illustration of these compensatory joint motions is shown in Figure 2-4. If the step width is identical, subjects with greater knee abduction seem to have compensated for smaller rearfoot eversion and greater hip adduction, and vice versa. Although static alignment was not measured in this study, Horton

and Hall (1989) reported that females had different static alignment. However, static measurements such as Q-angle are not always representative of joint kinematics during running (Heiderscheit et al., 2000; Willson and Davis, 2008a). Although there is a possibility that the subjects of the present study had different anatomical structural alignments, we consider that the gender-related differences in the frontal plane joint kinematics observed in this study were the result of this compensatory mechanism. It is speculated that because running is a closed kinetic chain exercise when the foot contacts the ground, alteration of step width may be beneficial to reduce knee abduction, since increasing the step width will result in less hip adduction, thereby reducing knee abduction.

Limitations and errors associated with the calculation of lower extremity kinematics have been well documented. For the rearfoot segment, external rearfoot markers were placed on the heel counter of the shoe in this study. Reinschmidt et al. (1997) reported that external markers on the shoe are only gross indicators of the skeletal tibiocalcaneal motion. Although there is a possibility that the ankle joint angles reported in this study overestimate the true tibiocalcaneal motion, it would have been consistent across all subjects because all subjects wore the same shoes during running.

Similarly, it is clear that tracking marker motions relative to the underlying bone might not represent an accurate estimation of segmental kinematics due to soft-tissue artifacts,

which is the most significant error in motion analysis (Leardini et al., 2005). In this study, we attempted to avoid areas of large muscle mass when attaching the tracking markers. The ICC values for the measured angles were greater than 0.86 for trial-retrial reliability and greater than 0.77 for test-retest reliability, indicating good to excellent reliability. In addition, the mean values of the RMS differences were less than 1.7°. These results indicate that the measured angles were reliable and the measurement errors were sufficiently smaller than the observed joint angles of interest in the present study. Thus, we expect that this soft-tissue artifact is systematic and does not significantly affect to the findings of this study.

2-5. Summary

The purpose of this study was to clarify gender differences in the lower extremity kinematics during running, with a specific emphasis on the relationships between the distal and proximal factors and knee joint kinematics. The current results indicate that if the step width is identical, the subjects with greater knee abduction should exhibit smaller rearfoot eversion to compensate for the greater hip adduction, which was more apparent in females. These relationships explain greater knee abduction found in female runners, which can be linked to a high risk of knee injury.
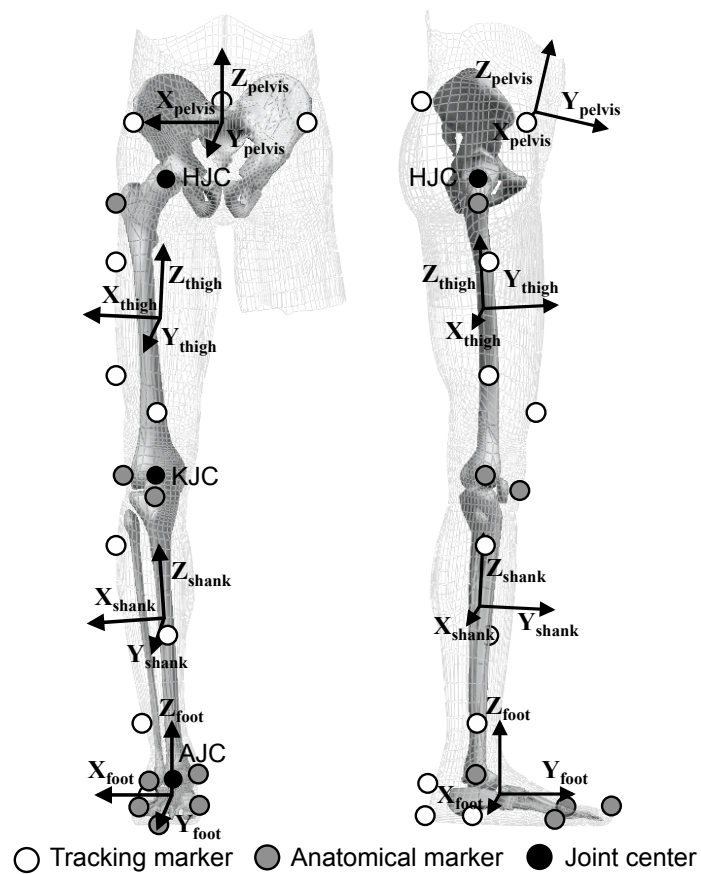
Figure 2-1. Frontal and lateral views of the marker locations and anatomical segment coordinate systems of the lower extremity.
Anatomical markers were used to define joint centers and to create anatomical segment coordinate systems while the tracking markers were used to track the position and orientation of the segments during the running trials.
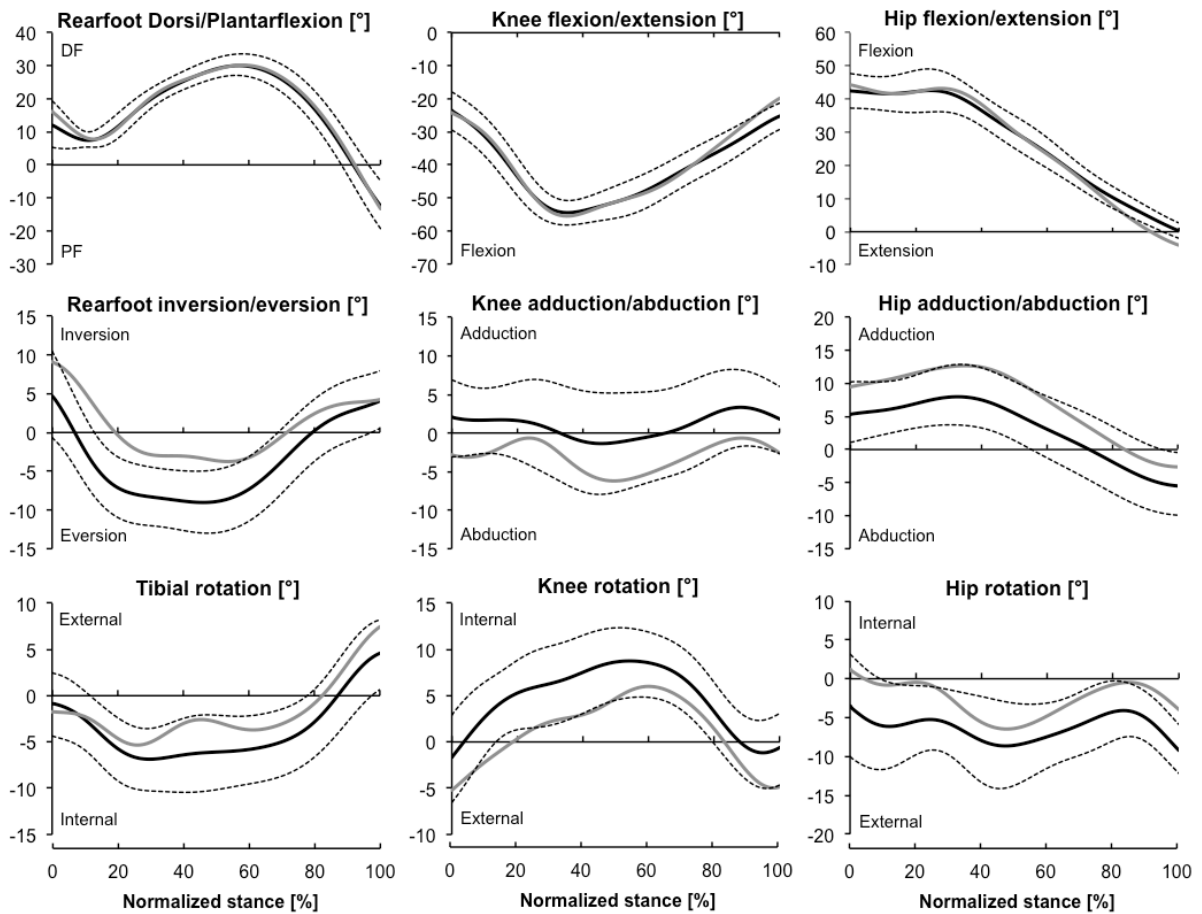
Figure 2-2. Three-dimensional joint angles for male (mean, solid black line; SD, dashed line) and female (mean, solid grey line) runners at the ankle, knee, and hip joints during the stance phase of running. Joint angles are in degrees.
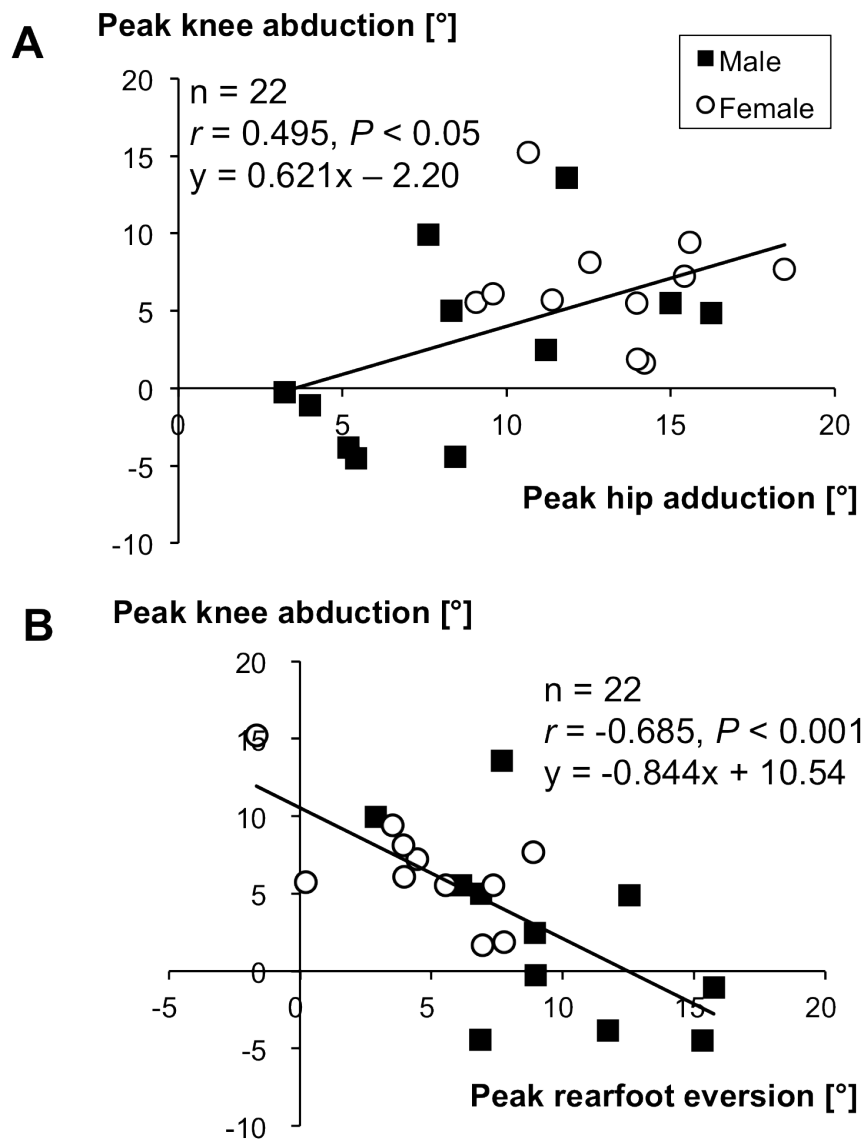
Figure 2-3. Pearson's correlations between peak knee abduction and (A) peak hip adduction, and (B) peak rearfoot eversion (filled squares are males, open circles are females).

Table 2-1. Mean values (± SD) of the peak joint angles, the instant at which peak joint angle was recorded at the ankle, knee and hip joints, and the normalized step width of both genders.

| Variables of interest	Female	Male	ES	P	
Peak joint angle [°]					
Rearfoot eversion	4.4 (3.4)	9.4 (3.9)	-1.37	0.002	**
Tibial internal rotation	6.9 (1.8)	7.9 (3.2)	-0.39	0.188	
Knee abduction	7.5 (4.3)	2.5 (5.9)	0.96	0.018	*
Knee internal rotation	6.7 (5.6)	9.5 (3.8)	-0.58	0.094	
Hip adduction	13.2 (3.1)	8.6 (4.2)	1.28	0.004	**
Hip internal rotation	4.7 (5.2)	-0.7 (5.5)	1.00	0.015	*
Peak timing [% stance]					
Rearfoot eversion	45.7 (9.9)	43.6 (8.3)	0.24	0.585	
Tibial internal rotation	36.5 (12.6)	37.6 (12.3)	-0.09	0.832	
Knee abduction	41.0 (17.3)	43.7 (10.2)	-0.19	0.663	
Knee internal rotation	51.0 (14.1)	48.9 (9.2)	0.17	0.687	
Hip adduction	29.2 (10.6)	30.3 (10.6)	-0.10	0.822	
Hip internal rotation	9.7 (7.9)	18.7 (13.4)	-0.81	0.071	
Normalized step width [%]	15.9 (8.6)	12.1 (6.1)	0.49	0.155	

*: $P<0.05$, **: $P<0.01$
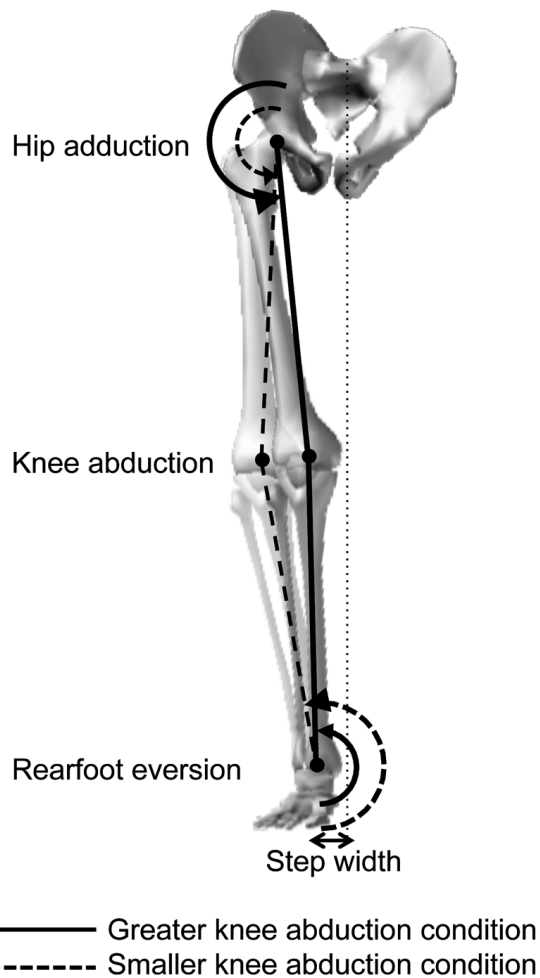
Figure 2-4. Schematic illustration of the compensatory motion in the frontal view. The solid lines represent the greater knee abduction condition and the dashed lines represent the smaller knee abduction condition. If the step width is identical, the subject with small rearfoot eversion demonstrates large knee abduction, similarly and the subject with large hip adduction demonstrates large knee abduction. These motions are typical in females.

Chapter 3   Relationships between ankle and hip joint kinematics and frontal

plane knee joint motions during running

3-1. Introduction

The knee is the most common site of running injury, accounting for 40% of all running injuries, and patellofemoral pain syndrome (PFPS) is reported as the most frequent running injury (Taunton et al., 2002). Large knee abduction angles during weight-bearing exercise have been reported as a kinematic risk factor of PFPS (Powers, 2003). Greater knee abduction increases the Q-angle during weight-bearing exercises and may result in lateral patellar dislocation and greater compressive stresses on the patellofemoral joint (Mizuno et al., 2000; Powers, 2003).

Recent studies have begun to explore the influence of ankle and hip joint kinematics on PFPS (Dierks et al., 2008; Davis and Powers, 2010; Noehren et al., 2012; Powers et al., 2012), with the rationale that, because the patella is anatomically linked to both the femur and tibia, the frontal and transverse plane motions of the ankle and hip could affect patellofemoral joint kinematics and contact pressure (Tiberio, 1987; Powers, 2003). Another reason why both ankle and hip joint kinematics could potentially affect knee joint kinematics is that there is little muscle around the knee that provides direct frontal plane control at the knee.

As a proximal factor, abnormal hip motions play crucial roles in the development of knee injuries (Powers, 2003; Dierks et al., 2008). The weakness of the hip abductor and external rotator muscles may increase hip adduction, internal rotation, and subsequent knee abduction (Ireland et al., 2003; Reiman et al., 2009). However, regarding hip internal rotation, inconsistent observations have been reported among studies that compared between subjects with PFPS and uninjured controls (Dierks et al., 2008; Willson and Davis, 2008a; Souza and Powers, 2009). Therefore, the influence of hip internal rotation on knee frontal plane kinematics is still unclear.

With respect to a distal factor, excessive foot pronation and/or rearfoot eversion have often been implicated in the development of running injuries at the knee (Tiberio, 1987; Hintermann and Nigg, 1998; Powers, 2003). Early studies reported that a high percentage of injured runners were over-pronators (James et al., 1978; Clement et al., 1981), and rearfoot kinematics have often been investigated as contributing factors to overuse running injuries (Messier et al., 1991; Duffey et al., 2000; Dierks et al., 2011; Noehren et al., 2012a). However, recent studies of three-dimensional analysis of rearfoot motion during running generally revealed that there was no significant difference in peak rearfoot eversion between injured and non-injured runners (Dierks et al., 2011; Noehren et al., 2012a). These studies suggest that the influence of rearfoot eversion on knee joint injuries is questionable.

Even though both ankle and hip joint kinematics have been known to relate with frontal plane knee joint kinematics, it is unclear which of these factors are strongly related to knee frontal plane motions. Therefore, the purpose of this study was to explore the relationships between ankle and hip joint kinematics and frontal plane knee joint kinematics during running. It was discovered in Chapter 2 that there were gender differences in the three lower-limb joint kinematics but the interrelationships between the hip and knee, and the ankle and knee, were similar for both genders. Thus, based on the previously proposed concepts (Tiberio, 1987; Powers, 2003) and the results of Chapter 2, it can be hypothesized that both greater ankle and hip joint motions in the frontal and transverse planes could be linked to greater knee abduction.

3-2. Methods

Subjects

Seventy runners (54 males and 16 females; mean age: 22.3 ± 3.6 years, height: 170.2 ± 7.2 cm, mass: 59.9 ± 7.5 kg, weekly running distance: 46.0 ± 34.3 km; mean ± SD) participated in this study. Inclusion criteria were that subjects had no lower extremity injury within 6 months prior to the measurement, and kept habitual running experience of at least 10 km/week. This study was approved by the Ethics Committee on Human Research of Waseda

University and before inclusion into the study all subjects were informed of the details of the

experiment as well as the purpose of the study and provided written informed consent.

Data Collection

During data collection, all subjects were fitted with the identical-model running

shoes (Adizero Boston, adidas AG, Germany) with moderate cushioning. Three markers per

segment were used to track the position and orientation of the pelvis, thigh, shank and foot

segments. Markers were attached to the following locations: sacrum which is the midpoint of

the two posterior superior iliac spine, right anterior superior iliac spine (ASIS), left ASIS,

proximal lateral thigh below the greater trochanter, distal lateral thigh, distal anterior thigh,

proximal lateral shank, midtibial crest, distal lateral shank, upper shoe heel, lower shoe heel,

and lateral side of the shoe below the lateral malleoli. When attaching the tracking markers,

the subjects were requested to contract the muscles of the right leg several times in the

standing position, and markers were placed to avoid areas of large muscle deformation.

Prior to the running trials, a standing neutral trial was recorded. In the neutral trial,

the subjects were asked to stand in a position with their feet approximately hip width apart,

with the knee and hip joints fully extended. For the neutral trial, additional anatomical markers

were placed on the following locations to define joint centers and anatomical segment

coordinate systems: greater trochanter, lateral femoral epicondyle, patella, medial and lateral malleoli, the heads of the 1st and 5th metatarsals and the tip of the shoe. After the neutral trial was recorded, the anatomical markers were removed.

Subjects then performed running trials on a 25-m runway at a speed of 4 m/s. The running speed was monitored with photocells (E3G-MR19T, Omron Corp., Tokyo, Japan) placed just before and after the force plate at a distance of 1.92 m apart. Each subject was asked to repeat the trial until the stance phases of ten successful trials were collected. A trial was judged successful if the subject ran within 5% of the target running speed and the right foot was in contact with the force plate with a natural running style. A motion capture system with eight infrared cameras (Motion Analysis Corp., Santa Rosa, CA) was used to determine three-dimensional marker positions sampling at 240 Hz. Ground reaction forces were simultaneously collected at 2400 Hz using a force plate (Bertec Corp., Columbus, OH) to determine heel-strike and toe-off. Before testing, cameras were calibrated to a defined capture volume (2.5m length × 1.2m width × 1.5m height), and the average three-dimensional residual was below 0.5 mm.

Data Reduction

All kinematic data were analyzed during the stance phase of running. Heel-strike

and toe-off were determined as the instants at which the vertical ground reaction force was greater or less than 20 N, respectively. The marker position data of the running trials were tracked for a period starting from 20 frames before to 20 frames after the stance phase. Custom-made Matlab software (R2011a, Mathworks Inc., Natick, MA) was used for data processing and analysis. Before calculation, the marker position data and ground reaction forces were filtered using a fourth-order zero lag Butterworth low-pass filter with a cut-off frequency of 12 Hz and 50 Hz, respectively (Park et al., 2009).

A right-handed orthogonal coordinate system embedded in each segment was defined by anatomical markers and joint centers during the neutral trial. Initially, the pelvis coordinate system was defined by the ASIS and sacrum markers (Wu et al., 2002) and then the hip joint center was estimated based on the inter-ASIS distance and pelvis coordinate system (Bell et al., 1989). The antero-posterior coordinate of the hip joint center was defined by the greater trochanter marker. The knee joint center was identified by markers on the lateral femoral epicondyle and the patella. The lateral femoral epicondyle marker defined the knee joint center's anterior-posterior and superior-inferior coordinates, and the patella marker defined the knee joint center's medio-lateral coordinate (Stefanyshyn et al., 2006). The midpoint of two markers on the lateral and medial malleoli was used to define the ankle joint center. Once joint centers were defined, the thigh and shank coordinate systems were defined

49

by the proximal and distal joint centers and lateral markers of the proximal joint of each

segment (greater trochanter for the thigh, lateral femoral epicondyle for the shank). The foot

coordinate system was aligned with the laboratory coordinate system, and then rotated around

the vertical axis so that its antero-posterior axis passed through the lower heel marker and the

tip of the shoe marker in the transverse plane. During the running trials, the singular value

decomposition method (Söderkvist and Wedin, 1993) was utilized to obtain the anatomical

segment coordinate systems.

Joint angles were determined as the orientation of the distal segment with respect to

the proximal segment according to the definition of the joint coordinate system (Grood and

Suntay, 1983; Cole et al., 1993). The ankle, knee and hip joint angles in the frontal and

transverse planes were selected as the kinematic variables of interest according to the

previously proposed concepts, specifically rearfoot eversion, tibial internal rotation, knee

abduction, hip adduction and hip internal rotation (Tiberio, 1987; Powers, 2003). The

time-series data of joint angles were time normalized to 101 data points, with each time

interval representing 1% of the stance phase. On visual inspection of the data, peak values of

variables of interest occurred between 20-80% of the stance phase. In addition, these variables

during the weight-bearing phase are considered to be more important. Therefore, peak values

of variables of interest were extracted from time-course data during mid-stance (20-80% of the

stance phase). Additionally, contralateral pelvic drop and thigh adduction angles at the instant

of peak hip adduction were analyzed. These segment angles were determined as the orientation

of each segment with respect to the laboratory coordinate system using a Cardan rotation

sequence of tilt, obliquity (abduction/adduction), and rotation.

Statistical Analyses

Pearson's correlation coefficients were used to determine the associations among the

variables of interest. A stepwise multiple regression analysis was conducted to identify which

ankle and hip joint kinematics associated with peak knee abduction. Based on the previously

proposed concept (Powers, 2003), the dependent variable was peak knee abduction, and the

independent variables were peak values of rearfoot eversion, tibial internal rotation, hip

adduction, and hip internal rotation angles. Statistical analyses were performed using SPSS

12.0 J (SPSS Inc., Chicago, IL). For all statistical analyses, the level of significance was set at

0.05.

3-3. Results

The ensemble average of joint angles in the frontal and transverse planes at the ankle,

knee and hip joints can be seen in Figure 3-1. The relationships among the variables of interest

are presented in Table 3-1. The peak knee abduction angle was positively correlated with hip

adduction angle ($r = 0.283$, $P < 0.05$), and negatively correlated with peak rearfoot eversion (r

$= -0.638$, $P < 0.01$), peak tibial internal rotation angle ($r = -0.274$, $P < 0.05$) and hip internal

rotation angles ($r = -0.700$, $P < 0.01$). Peak rearfoot eversion was also positively correlated

with peak tibial internal rotation ($r = 0.442$, $P < 0.01$). The stepwise multiple regression

analysis revealed that peak hip internal rotation, rearfoot eversion and hip adduction angles

were significant predictors for the knee abduction angles in the order of the standardized

partial correlation coefficients (coefficient of determination: $R^2 = 0.737$, $P < 0.001$; Table 3-2).

Additionally, the peak hip adduction angle was positively correlated with contralateral pelvic

drop ($r = 0.623$, $P < 0.01$) and thigh adduction angles ($r = 0.658$, $P < 0.01$) at the instant of

peak hip adduction angle. There was no significant correlation between contralateral pelvic

drop and thigh adduction angles at the instant of peak hip adduction angle ($r = 0.179$, $P >$

0.05).

3-4. Discussion

The present study aimed to establish how ankle and hip joint kinematics associate

with frontal plane knee joint kinematics during running. The results of this study showed that

ankle and hip joint kinematics were significantly associated with knee frontal plane motion, as

hypothesized. Peak hip internal rotation, rearfoot eversion and hip adduction angles were selected as the significant predictors for knee abduction angle. This result indicates that subjects with the combination of smaller hip internal rotation, smaller rearfoot eversion and greater hip adduction tend to demonstrate greater knee abduction, and vice versa. Therefore, the results of this study contradict the proposed associations among ankle, knee and hip joint kinematics (Tiberio, 1987; Powers, 2003), especially regarding hip rotation and rearfoot eversion.

Hip internal rotation angle was identified as the strongest predictor of frontal plane knee joint motions. Contrary to the current theoretical concept for medial collapse mechanism of PFPS that has focused on the influence of increased hip internal rotation (Powers, 2003), the results of this study indicate that subjects who tended to have smaller hip internal rotation (or greater hip external rotation) demonstrated greater knee abduction. In addition, findings in the previous studies that compared between subjects with PFPS and uninjured controls do not consistently support the previously proposed concept of PFPS regarding hip internal rotation (Dierks et al., 2008; Willson and Davis, 2008a; Souza and Powers, 2009). Indeed, increased hip external rotation or decreased hip/femur internal rotation was identified in symptomatic subjects in non-running activities, such as walking, single leg jump and single leg squat (Powers et al., 2002; Willson and Davis, 2008a, 2009). Although Willson and Davis (2008a)

postulated that smaller hip internal rotation was a compensation mechanism employed to decrease retropatellar stress and pain, the results of the current study have shown that the subjects who had smaller hip internal rotation angles demonstrated greater knee abduction angles. Therefore, smaller hip internal rotation may not be a compensatory motion for decreasing stress and pain, but may play a crucial role in the development of knee injuries.

Rearfoot eversion was also associated with frontal plane knee joint motion. Due to a theoretical linking with tibial rotation, rearfoot eversion is frequently suggested as a risk factor of running injuries at the knee because the patella is anatomically attached to the tibia (Tiberio, 1987; Powers, 2003; Ferber et al., 2009). In fact, rearfoot eversion was positively correlated with tibial internal rotation in this study. However, the results of this study revealed that subjects with smaller rearfoot eversion demonstrated greater knee abduction. Thus, it would appear that there also exists a different relationship between ankle joint kinematics and knee frontal plane motions opposing previously proposed concepts (Tiberio, 1987; Powers, 2003).

The only variable that supported the previously proposed concepts was hip adduction. In fact, several studies commonly reported that subjects with PFPS demonstrated greater hip adduction than asymptomatic control subjects (Willson and Davis, 2008a; Noehren et al., 2012). Weaker hip abductor strength has been suggested to contribute to an increase in hip adduction and contralateral pelvic drop (Ireland et al., 2003; Leetun et al., 2004; Niemuth

et al., 2005; Cichanowski et al., 2007). Leetun et al. (2004) prospectively compared differences in hip strength, and reported that injured athletes showed significantly weaker hip abductor strength than uninjured athletes. Nevertheless, the correlation coefficient between peak hip adduction and frontal plane knee joint motion was not high. Hip adduction angle would be affected by not only thigh segment frontal plane motion but by the contralateral pelvic drop; if thigh segment frontal plane motion does not change, hip adduction angle can be changed due to contralateral pelvic drop. In fact, both thigh adduction and contralateral pelvic drop angles at the instant of peak hip adduction angle were correlated with peak hip adduction angle. However, thigh adduction angle was not correlated with contralateral pelvic drop. This might be why the correlation coefficient between hip adduction and frontal plane knee motion was low.

The relationship between lower extremity kinematics and knee frontal plane motion may be explained by taking the external knee adduction moment into account. Willson and Davis (2008a) reported that subjects with PFPS displayed smaller hip internal rotation and an average of 5° greater toe-out angle compared to asymptomatic controls during running. Thus, smaller hip internal rotation may be related with toe-out angle during the stance phase of running. Moreover, greater toe-out angle has also been reported as a strategy to reduce external knee adduction moment, the moment of the ground reaction forces and inertial forces about the

joint center (Andrews et al., 1996; Jenkyn et al., 2008), which is balanced by the internal knee

abduction moment generated by muscles, soft tissues and joint contact forces. These findings

suggest that increasing hip external rotation angle may result in greater toe-out angle. This

movement would decrease external knee adduction moment, and decreased external knee

adduction moment may also result in decreased shank adduction angle. Since the shank

segment is a segment common to both ankle and knee joints, if the adduction of this segment

is small, rearfoot eversion angle would be small and knee abduction angle would be large.

Future research is needed to clarify these mechanisms.

One limitation of this study is the pooling of data from both male and female runners.

It is well known that there are differences in running kinematics between genders (Ferber et al.,

2003; Chumanov et al., 2008). A common observation among these studies is that female

runners demonstrate greater knee abduction, greater hip adduction as well as greater hip

internal rotation than male runners. However, it was discovered in Chapter 2 that the

interrelationships between the hip and knee, and the ankle and knee, were similar for both

genders, and so we did not focus on gender-specific differences in running kinematics but

intended to investigate the relationships among ankle, knee and hip joint kinematics in a

healthy population including both males and females with a wide range of data. In addition,

pooling the data increased the power of analysis and allowed us to simplify the applicability of

our data to both populations.

3-5. Summary

The purpose of this study was to establish how ankle and hip joint kinematics are associated with frontal plane knee joint kinematics during running. The findings of this study indicate that hip internal rotation, rearfoot eversion and hip adduction angles are related to knee abduction, a kinematic risk factor of running injury at the knee. There appear to exist a mechanism that differs from previously suggested concepts. The results of this study showed that smaller hip internal rotation, smaller rearfoot eversion and greater hip adduction were found to be associated with greater knee abduction.
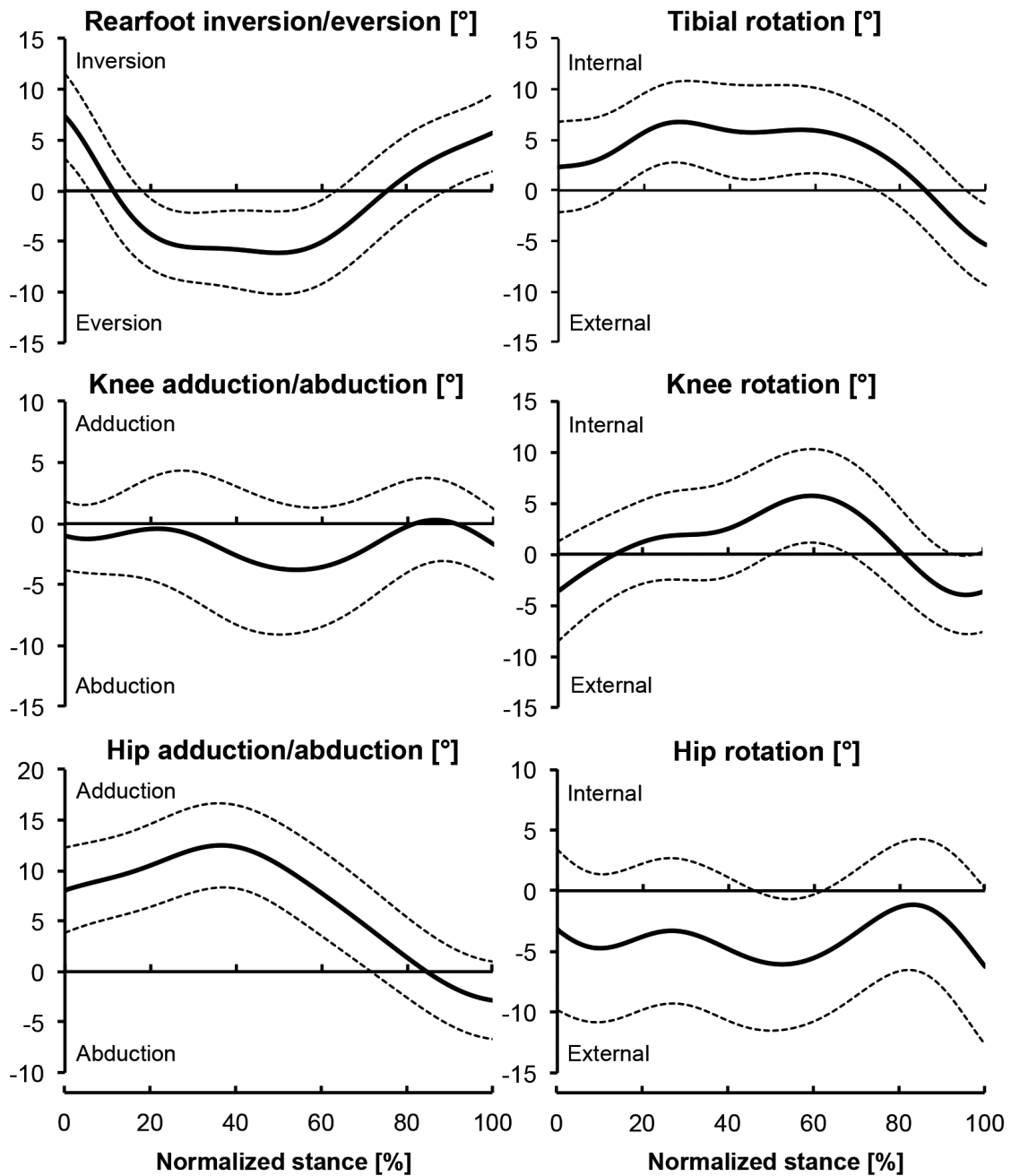
Figure 3-1. The ensemble average of joint angles in the frontal and transverse planes (mean, solid black line; SD, dashed line) at the ankle, knee and hip joints during the stance phase of running. Joint angles are in degrees.

Table 3-1. Correlation matrix for the kinematic variables of interest.

	Rearfoot eversion		Tibial int. rot.		Knee abduction		Hip adduction
Tibial internal rotation	0.442	**					
Knee abduction	-0.638	**	-0.274	*			
Hip adduction	-0.080		-0.036		0.283	*	
Hip internal rotation	0.329	**	0.126		-0.700	**	0.001

**: $P < 0.01$, *: $P < 0.05$

Table 3-2. The results of stepwise multiple regression analysis.

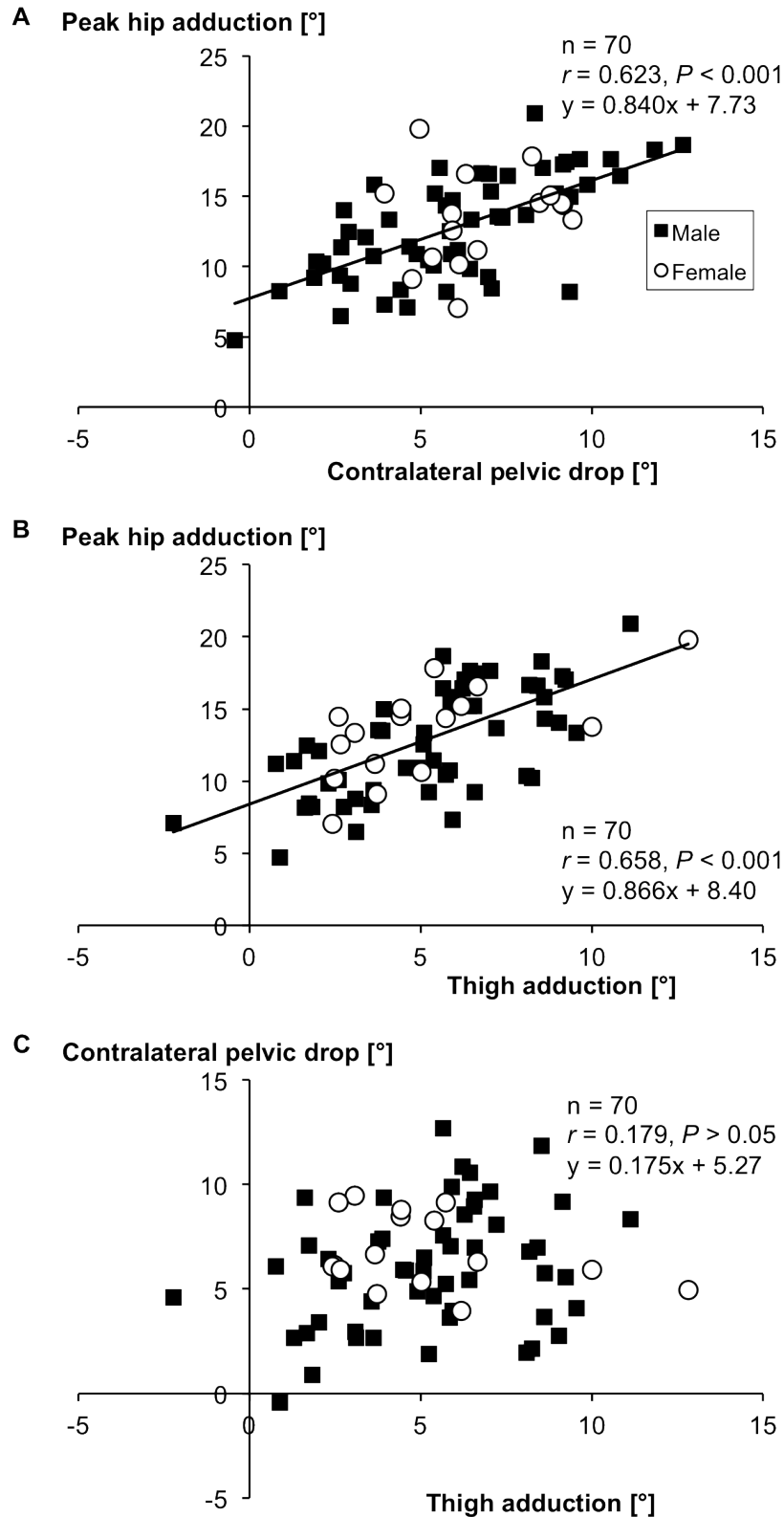| Dependent varible	Independent variable	β	*P*
Knee abduction	Hip internal rotation	-0.557	< 0.001
	Rearfoot eversion	-0.434	< 0.001
	Hip adduction	0.248	< 0.001

β: Standardized partial correlation coefficient

Figure 3-2. Relationships between (A) peak hip adduction and contralateral pelvic drop at peak hip adduction and (B) thigh adduction at peak hip adduction, and (C) the relationship between contralateral pelvic drop at peak hip adduction and thigh adduction at peak hip adduction (filled squares are males, open circles are females).

61

Chapter 4   **Hip rotation angle is associated with frontal plane knee joint loading**

during running

4-1. Introduction

The knee is the most common site of running injury, accounting for 40% of all running injuries (Taunton et al., 2002). It has been proposed that inability to control lower extremity segments (Tiberio, 1987; Ireland et al., 2003; Powers et al., 2003) and knee joint loads (Stefanyshyn et al., 2006) in the frontal and transverse planes may associate with running injuries at the knee. Recently, inadequate hip abduction and external rotation strength has been suggested to influence the control of hip joint motions and subsequent knee injuries. Specifically, greater hip adduction and hip internal rotation angles are believed to relate to running injuries (Ireland et al., 2003; Powers et al., 2003). However, inconsistent observations regarding hip internal rotation angles have been reported among studies that compared between subjects with patellofemoral pain syndrome (PFPS) and uninjured controls during running (Dierks et al., 2008; Willson and Davis, 2008a; 2008b; Souza and Powers, 2009). Thus, although large hip internal rotation is generally considered to play a role in the development of running injuries at the knee (Ireland et al., 2003; Powers et al., 2003), the influence of hip rotation angles on knee frontal plane kinematics and kinetics remains unclear.

With respect to the knee joint loading, Stefanyshyn et al. (2006) have shown that runners with PFPS exhibit greater internal knee abduction impulses, which are quantified by integrating the internal knee abduction moment–time curve, than asymptomatic healthy runners in both retrospective and prospective studies, and increased knee abduction impulse is predictive of developing PFPS. Increased toe-out angle has been suggested as a strategy to reduce the external knee adduction moment (Andrews et al., 1996; Jenkyn et al., 2008), the moment of the ground reaction forces and inertial forces about the joint center, which is balanced by the internal knee abduction moment generated by muscles, soft tissues and joint contact forces. The knee joint moment in the frontal plane is related to the location of the medio-lateral center of pressure. For example, medial and lateral wedged footwear are frequently investigated as a method to reduce frontal plane moments at the knee during walking and running (Willams et al., 2003; Hinman et al., 2008; Russell et al., 2011). It has been reported that lateral wedges significantly reduce internal knee abduction moments by shifting the center of pressure location laterally (Fisher et al., 2007; Russell et al., 2011). Because center of pressure location is located under the foot during the stance phase of running, toe-out angle may also be related to knee joint loading in the frontal plane. Therefore, an assumed mechanism to reduce knee abduction moment can be that greater toe-out angle would shift the center of pressure laterally and move the line of action of the ground reaction

force (GRF) closer to the knee joint center, thus reducing the frontal plane lever arm at the knee.

On the other hand, large knee abduction angle during weight-bearing exercise has also been considered as a kinematic risk factor of PFPS (Powers, 2003). Moreover, the studies of Willson and Davis (2008a; 2008b) reported that subjects with PFPS displayed smaller hip internal rotation, smaller knee adduction and an average of 5° greater toe-out angle compared to asymptomatic controls during running. Taking into account the assumed mechanism to reduce frontal plane knee joint moment as mentioned above, increasing hip external rotation would result in greater toe-out angle, and this movement would decrease the internal knee abduction moment. However, if so, the kinematic and kinetic risk factors of PFPS may not be corresponding.

Consequently, the purpose of this study was to explore how hip transverse plane motions are related with frontal plane knee joint loading associated with the risk factor of running injury at the knee. It was hypothesized that hip rotation angle would be related with toe-out angle, and greater toe-out angle would result in a shorter frontal plane lever arm since center of pressure location would shift laterally.

4-2. Methods

Subjects

Seventy runners (54 males and 16 females; mean age: 22.3 ± 3.6 years, height: 170.2 ± 7.2 cm, mass: 59.9 ± 7.5 kg, weekly running distance: 46.0 ± 34.3 km; mean ± SD) participated in this study. Inclusion criteria were that subjects kept habitual running experience of at least 10 km/week and had been free from injury of the lower extremity within 6 months prior to the measurement. This study was approved by the Ethics Committee on Human Research of Waseda University and informed written consent was obtained from all subjects prior to the experiment.

Data Collection

Three-dimensional kinematics and kinetics of the lower limb were collected for the right leg of each subject. All subjects were fitted with the identical-model running shoes (Adizero Boston, adidas AG, Germany) during data collection. Three-dimensional reflective marker positions and GRF were simultaneously recorded with an 8-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA, USA) and a force plate (Bertec Corporation, Columbus, OH, USA) sampling at 240 Hz and 2400 Hz, respectively. Before testing, cameras were calibrated to a defined capture volume (2.5m length × 1.2m width ×

1.5m height), and the average three-dimensional residual was below 0.5 mm. Three retro-reflective markers per segment were used to track the position and orientation of the segments, attached to the following locations: sacrum which is the midpoint of the two posterior superior iliac spine, right anterior superior iliac spine (ASIS), left ASIS, proximal lateral thigh below the greater trochanter, distal lateral thigh, distal anterior thigh, proximal lateral shank, midtibial crest, distal lateral shank, upper shoe heel, lower shoe heel, and lateral side of the shoe below the lateral malleoli. When attaching the tracking markers, the subjects were requested to contract the muscles of the right leg several times in the standing position, and we attempted to avoid the areas of large muscle deformation.

Before starting the running trials, a standing neutral trial was recorded to determine the tracking marker positions relative to the anatomical segment coordinate systems and joint centers for each segment. In the neutral trial, the subjects were asked to stand in a position with their feet approximately hip-width apart, with the knee and hip joints fully extended. In addition, the long axis of the foot segment was aligned with the antero-posterior axis of the laboratory coordinate system as closely as possible. For this trial, additional anatomical markers were placed on the following locations to define joint centers and anatomical segment coordinate systems: greater trochanter, lateral femoral epicondyle, patella, medial and lateral malleoli, the heads of the 1st and 5th metatarsals and the tip of the shoe. After the standing

neutral was collected, these anatomical markers were removed.

Subjects then performed running trials on a 25-m runway at a speed of 4 m/s. Subjects were given several practice trials to ensure that the foot landed with a natural running style on the force plate. The running speed was monitored using photocells (E3G-MR19T, Omron Corp., Tokyo, Japan) placed just before and after the force plate at a distance of 1.92 m apart. Ten successful trials were collected during the stance phase of running. A successful trial was identified if the subject's right foot was in contact with the force plate correctly with a natural running style and the running speed was within 5% of the predefined speed.

Data Reduction

All kinematic and kinetic data were analyzed for the stance phase of running. Heel-strike was determined when the vertical GRF rose past 20N, and toe-off was determined when the vertical GRF fell below 20 N. The marker position data of the running trials were tracked for a period corresponding to 20 frames before and after contact with the force plate. Custom-made Matlab software (R2011a, Mathworks Inc., Natick, MA) was used for data processing and analysis. Before calculation, the marker position data and GRF data were filtered using a fourth-order zero lag Butterworth low-pass filter with a cut-off frequency of 12 Hz and 50 Hz, respectively.

A right-handed orthogonal coordinate system embedded in each segment was defined by anatomical markers and joint centers in a neutral trial. The pelvis coordinate system was defined by the ASIS and sacrum markers (Wu et al., 2002). The thigh and shank coordinate systems were defined by the proximal and distal joint centers and lateral markers of the proximal joint of each segment (greater trochanter for the thigh, lateral femoral epicondyle for the shank). The foot coordinate system was aligned with the laboratory coordinate system, and then rotated around the vertical axis so that its antero-posterior axis passed through the lower heel marker and the tip of the shoe marker in the transverse plane. The method of Söderkvist and Wedin (1993) was utilized to obtain the anatomical segment coordinate systems during running trials.

Joint angles were determined as the orientation of the distal segment with respect to the proximal segment according to the definition of the joint coordinate system (Grood and Suntay, 1983; Cole et al., 1993). In addition, foot segment angles with respect to the laboratory coordinate system were also determined according to a previous study (Areblad et al., 1990). Joint moments were calculated with a Newton-Euler inverse dynamics approach (Vaughan et al., 1992). The inertial properties of the segments used in the inverse dynamics calculations were estimated based on the subjects' mass and height (Ae et al., 1992). The joint moments were expressed in the joint coordinate system and represented the internal load on the joint,

which is balanced by the external moment created by ground reaction forces and inertial forces about the joint center (Stefanyshyn et al., 2006). The knee abduction impulse was quantified by integrating the internal knee abduction moment-time curve and represents the cumulative load during the stance phase of running (Stefanyshyn et al., 2006; Park and Stefanyshyn, 2011).

The knee joint moment is largely affected by the resultant GRF and the perpendicular distance from the line of action of the resultant GRF to the knee joint center. The lever arm of the GRF in the frontal plane of the shank segment about the knee joint center, which is termed as the frontal plane lever arm, was calculated according to the method described in the previous studies (Hunt et al., 2006; Jenkyn et al., 2008). In addition, the medio-lateral distance from the ankle joint center to the center of pressure (AJC-CoP) was calculated. Knee joint moment and impulse were normalized to body mass and height (Willson and Davis, 2008), and ground reaction forces were normalized to body weight. As well, frontal plane lever arm and AJC-CoP were normalized to body height.

The coefficient of determination in the relationship between knee abduction impulse and knee abduction moment at peak vertical GRF ($R^2 = 0.929$, $P < 0.001$; Figure 4-1 A), explaining 93% of the variance, was much higher than that between knee abduction impulse and peak knee abduction moment ($R^2 = 0.648$, $P < 0.001$; Figure 4-1 B). Thus, kinematic and

kinetic discrete variables of interest were reported at the instant of peak vertical GRF during

running. Pearson's correlation coefficients were used to determine the associations among the

knee abduction moment, frontal plane lever arm, medio-lateral and frontal plane GRF and hip

rotation and toe-out angles. Statistical analyses were performed using SPSS 12.0 J (SPSS Inc.,

Chicago, IL). For all statistical analyses, the level of significance was set at 0.05.

4-3. Results

The ensemble average of medio-lateral and vertical components of GRF, frontal

plane knee joint angle and moment, hip rotation angle and foot rotation angle can be seen in

Figure 4-2. A significant positive correlation existed between frontal plane lever arm and knee

abduction moment at peak vertical GRF ($r = 0.765$, $P < 0.001$; Figure 4-3 A). No significant

correlation was found between knee abduction moment at peak vertical GRF and either

medio-lateral ($r = -0.144$, $P > 0.05$; Figure 4-3 B) or vertical GRF ($r = 0.066$, $P > 0.05$; Figure

4-3 C). In addition, frontal plane lever arm was correlated positively with hip rotation angle

and negatively with toe-out angle and AJC-CoP. Toe-out angle was positively correlated with

AJC-CoP, and both toe-out angle and AJC-CoP were negatively correlated with hip rotation

angle (Table 4-1).

70

4-4. Discussion

The purpose of this study was to explore the relationships between lower extremity kinematics and frontal plane knee joint loading during running. The results of this study supported the stated hypotheses. Hip rotation angle was associated with toe-out angle, frontal plane lever arm and AJC-CoP distance.

Frontal plane lever arm was positively correlated with knee abduction moment at peak vertical GRF. However, neither medial lateral nor vertical GRF was correlated with knee abduction moment. This result indicates that although knee abduction moment is primarily calculated as the product of the resultant GRF in the frontal plane and the frontal plane lever arm, frontal plane lever arm has a greater influence on knee abduction moment. McClay and Manal (1999) reported that the variability of vertical GRF was relatively small, in contrast, the variability of medio-lateral GRF was relatively high but the magnitude was small compared to the antero-posterior and vertical components of GRF. These results indicate that the configurations of lower extremity joints may have a large influence on the lever arm of the GRF around the knee joint center in the frontal plane.

Frontal plane lever arm was correlated positively with hip internal rotation angle and negatively with toe-out angle and AJC-CoP. Additionally, toe-out angle was correlated positively with AJC-CoP distance and negatively with hip internal rotation angle. Collectively,

these results can be interpreted that subjects with smaller hip internal rotation/greater hip external rotation angles resulted in greater toe-out angles, greater AJC-CoP distances, and subsequent smaller frontal plane lever arms. Internal knee abduction moment/impulse have been suggested as a risk factor of PFPS (Stefanyshyn et al., 2006). Therefore, to decrease frontal plane lever arm, greater hip external rotation would be beneficial.

On the other hand, greater knee abduction angle has also been considered as a risk factor of PFPS (Powers et al., 2003). In Chapter 3, we found that greater hip external rotation angle was associated with greater knee abduction angle. Taking into consideration this result, there appears to be a conflict between kinematic (greater knee abduction angle) and kinetic (greater knee abduction impulse) risk factors of PFPS during running. This apparently conflicting result can be explained by the following mechanism which may be a possible reason why the kinematic and kinetic risk factors of PFPS were negatively correlated; greater hip external rotation angle would result in greater toe-out angle, and this movement would decrease knee abduction moment/impulse and shank adduction, and, as a consequence, increase knee abduction angle. However, this apparently conflicting result can not be explained perfectly by this mechanism, because the magnitude of the joint moment does not necessarily correspond to the changes in joint angle during dynamic activities. Further studies are needed to clarify this discrepancy between the kinematic and kinetic risk factors of PFPS.

Recently, inadequate hip abduction and external rotation strength has been considered to affect abnormal hip joint motions, and inability to control lower extremity segments in the frontal and transverse planes has been considered to play a crucial role in the development of knee injuries (Ireland et al., 2003; Reiman et al., 2009). Thus, it has been postulated that improving hip abduction and external rotation strength may stabilize hip frontal and transverse plane motions and reduce the risk of running injuries at the knee. Snyder et al. (2009) investigated the effect of 6-weeks of hip resistance training on lower extremity joint kinematics and kinetics during running, and reported that increased hip muscle strength altered the joint angles and moments. However, whereas the internal knee abduction moment decreased by 10%, the changes in joint range of motion, which significantly reduced following resistance training, were below 2°. Since the resistance training may be hard and its effect on the joint kinematics may be small, alteration of running style, such as an increased toe-out angle coupled with hip external rotation, would be beneficial to change knee joint mechanics.

4-5. Summary

The purpose of this study was to explore how hip rotation angles are related with frontal plane knee joint loading during running. The findings of this study indicate that frontal plane lever arm has a large influence on the knee abduction moment, and that greater hip

external rotation angles resulted in greater toe-out angles, greater AJC-CoP distances, and

subsequent smaller frontal plane lever arms. Therefore, to decrease the distance of the frontal

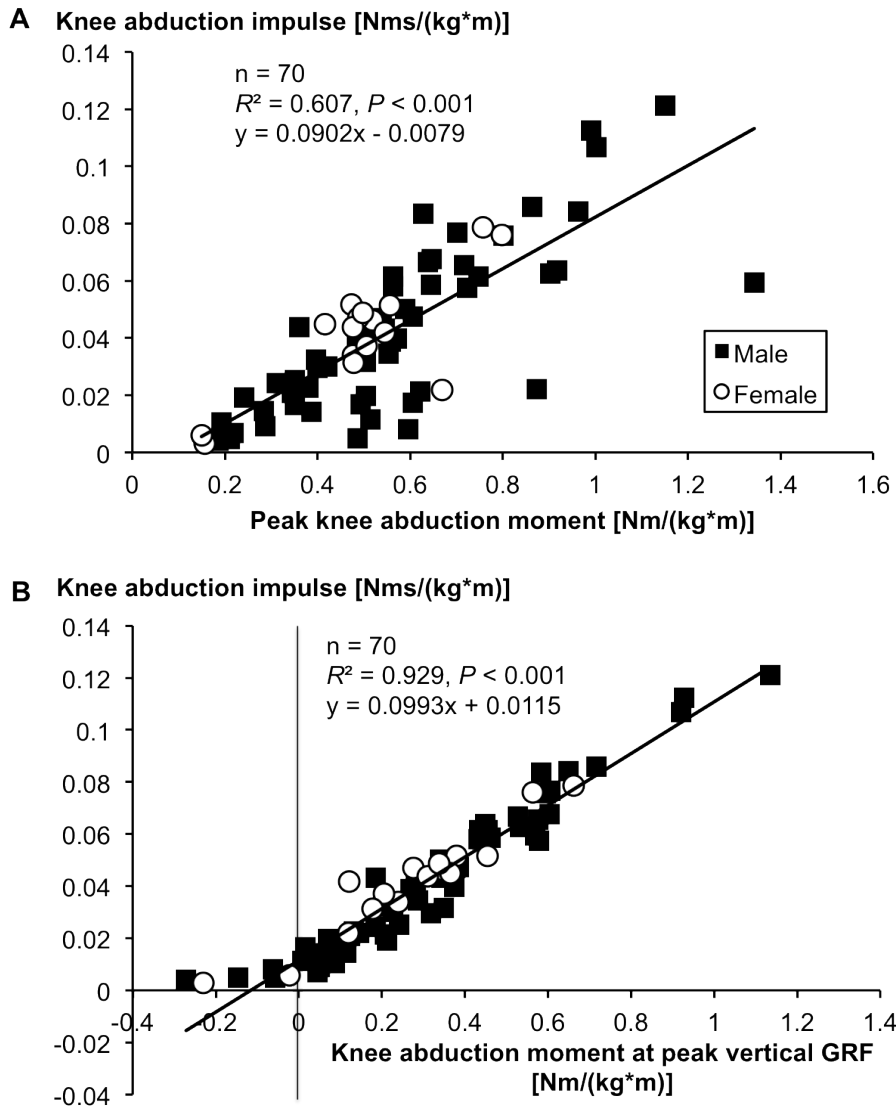plane lever arm, greater hip external rotation would be beneficial.

Figure 4-1. Relationships between knee abduction impulse and (A) peak knee abduction moment and (B) knee abduction moment at peak vertical GRF (filled squares are males, open circles are females).
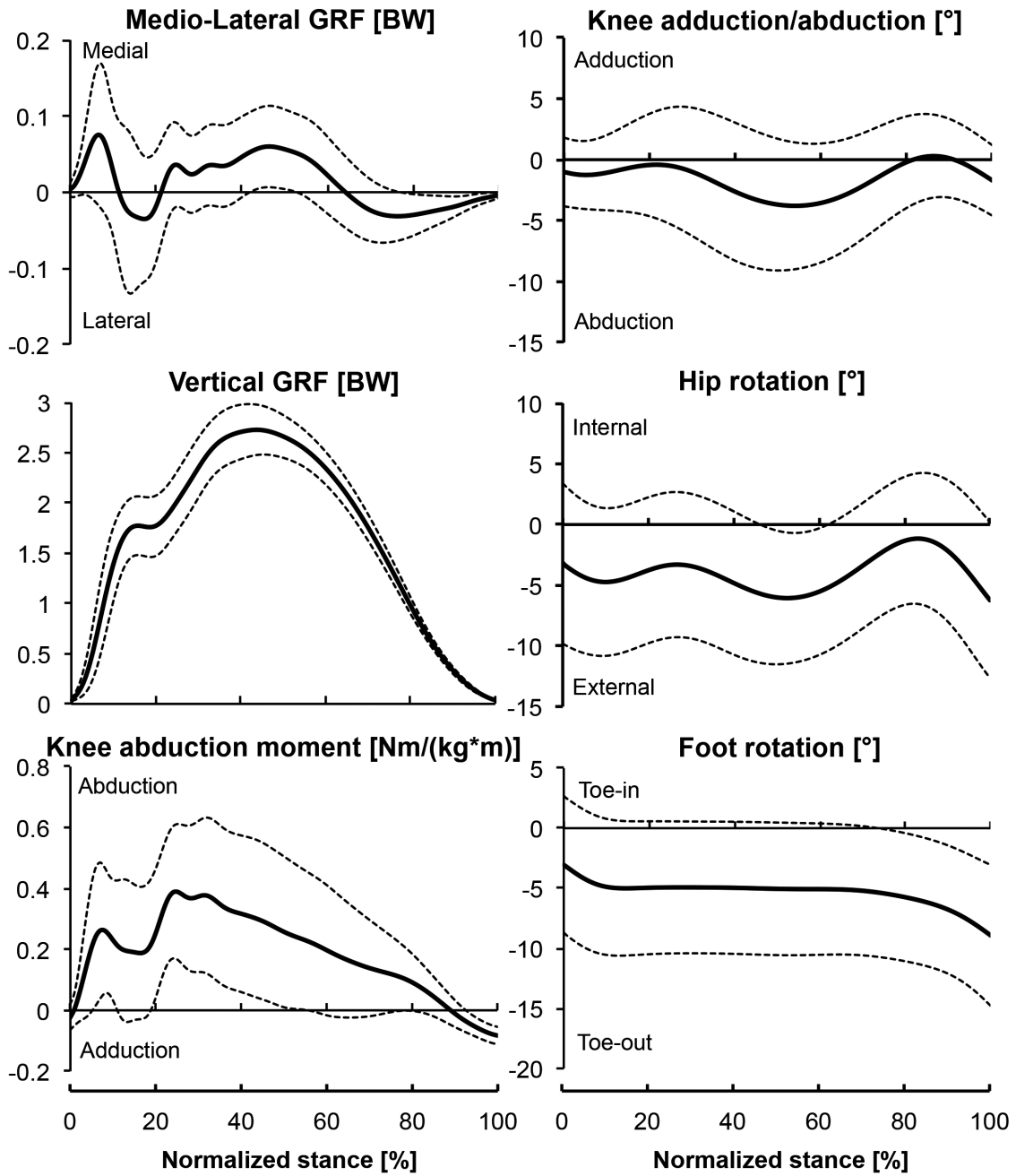
Figure 4-2. The ensemble average of medio-lateral GRF, vertical GRF, knee abduction moment, knee adduction/abduction angle, hip rotation angle and foot rotation angle (mean, solid black line; SD, dashed line) during the stance phase of running.
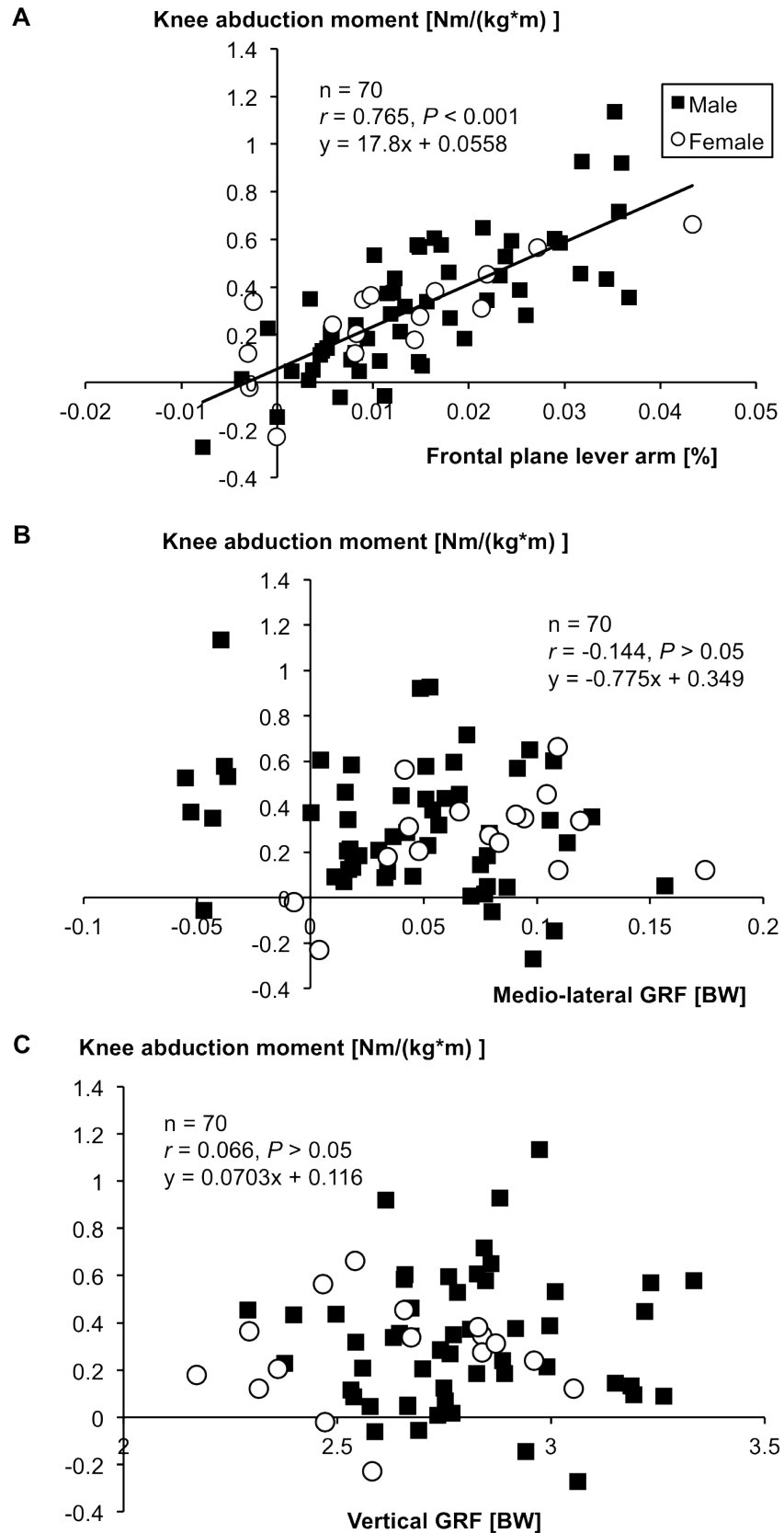
A

Knee abduction moment [Nm/(kg*m)]

n = 70
r = 0.765, *P* < 0.001
y = 17.8x + 0.0558

■ Male
○ Female

Frontal plane lever arm [%]

B

Knee abduction moment [Nm/(kg*m)]

n = 70
r = -0.144, *P* > 0.05
y = -0.775x + 0.349

Medio-lateral GRF [BW]

C

Knee abduction moment [Nm/(kg*m)]

n = 70
r = 0.066, *P* > 0.05
y = 0.0703x + 0.116

Vertical GRF [BW]

Figure 4-3. Relationships between knee abduction moment at peak vertical GRF and (A) frontal plane lever arm at peak vertical GRF, (B) medio-lateral GRF at peak vertical GRF and (C) peak vertical GRF (filled squares are males, open circles are females).

Table 4-1. Correlation matrix for the kinematic variables of interest.

	Frontal plane lever arm	Toe-out angle	AJC-CoP
Toe-out angle	-0.469 *		
AJC-CoP	-0.421 *	0.746 *	
Hip rotation	0.454 *	-0.465 *	-0.473 *

*: $P < 0.01$

Chapter 5   General discussion

In this chapter, the main findings of each previous chapter are first described. Then, the influences of ankle and hip joint kinematics on both kinematic and kinetic risk factors of running injury at the knee are discussed to elucidate possible mechanisms. Lastly, several factors that might affect the interpretation of the present results are addressed.

5-1. Main findings of each chapter

Running is one of the most common exercises because of its low cost, convenience and health benefits. However, the overall incidence rate of running injuries has been estimated at 29-56% per year. The most common site of running injury is the knee, accounting for approximately 40% of running injuries. Above all, PFPS prevails on running populations. A number of studies have linked running injuries to biomechanical variables, such as running kinematics and kinetics. Recently, both ankle and hip joint kinematics in the frontal and transverse planes are considered to influence frontal plane knee joint mechanics that has shown to influence PFPS. They can potentially lead to knee injuries during a closed kinetic chain exercise such as running, because, anatomically, there are limited adductor and abductor muscles around the knee. As a proximal factor, abnormal hip mechanics plays a crucial role in

the development of knee injuries (Powers, 2003; Dierks et al., 2008). The weakness of the hip abductor and external rotator muscles may increase hip adduction, internal rotation, and subsequent knee abduction (Ireland et al., 2003). As a distal factor, excessive foot pronation (eversion) has often been implicated in the development of running injuries at the knee (Tiberio, 1987; Hintermann and Nigg, 1998; Powers, 2003). Even though both ankle and hip joint kinematics have been related to knee joint mechanics, it is unclear how these factors are related. Therefore, the purpose of this thesis was to elucidate possible mechanisms how ankle and hip joint kinematics associate with frontal plane knee joint mechanics during running.

Main findings of each chapter and interpretations are summarized as follows:

Chapter 2: Female runners demonstrated smaller rearfoot eversion, greater knee abduction, greater hip adduction and greater hip internal rotation than male runners. There was a significant positive correlation between peak knee abduction and peak hip adduction, while there was a significant negative correlation between peak knee abduction and peak rearfoot eversion. These results indicate that if step width is identical, subjects with greater knee abduction have smaller rearfoot eversion to compensate for the greater hip adduction, which were more apparent in females. These relationships may explain the greater knee abduction found in female runners, which can be linked to a high risk of knee injury.

Chapter 3: The findings of Chapter 3 indicate that hip rotation, rearfoot eversion and hip adduction angles are related to the kinematic risk factor of running injury at the knee. However, a mechanism different than what has been shown previously appears to exist as greater hip external rotation, smaller rearfoot eversion and greater hip adduction were found to be associated with greater knee abduction.

Chapter 4: The findings of Chapter 4 indicate that the frontal plane lever arm of the knee joint moment has greater influence on the knee abduction moment than ground reaction forces. Also, subjects with smaller frontal plane lever arms demonstrated that greater hip external rotation angles resulted in greater toe-out angles, greater AJC-CoP distances, and subsequent smaller internal knee abduction moments.

5-2. The relationships between ankle and hip joint kinematics and frontal plane knee joint mechanics

Based on the results of Chapter 3 and Chapter 4, the relationships between knee abduction angle and frontal plane knee joint impulse can be seen in Figure 5-1 and 5-2. Knee abduction angles were negatively correlated with internal knee abduction moment and impulse. These results indicate that the subjects with greater internal knee abduction impulse (kinetic

factor believed to be an increased risk of PFPS) demonstrated smaller knee abduction angle (kinematic factor believed to be a decreased risk of PFPS), and vice versa, regardless of gender. Thus, there appears to be a conflict between kinematic and kinetic risk factors of PFPS during running. This apparently conflicting result can be explained by the following mechanism which may be a possible reason why the kinematic and kinetic risk factors of PFPS were negatively correlated; greater hip external rotation angle would result in greater toe-out angle, and this movement would decrease knee abduction moment/impulse and shank adduction. Since the shank segment is a segment common to both ankle and knee joints, if the adduction of this segment is small, knee abduction angle would be large and rearfoot eversion angle would be small (Figure 5-3). However, this apparently conflicting result can not be explained perfectly by this mechanism, because the magnitude of the joint moment does not necessarily correspond to the changes in joint angle during dynamic activities.

More recently, since male subjects with PFPS demonstrated greater peak knee adduction angle and external knee adduction moment compared with females with PFPS and with healthy males, there may exist a different mechanism between genders in the development of PFPS (Willy et al., 2012). The mean values of frontal plane knee joint angles, internal knee abduction impulse and moment are summarized in Figure 5-4 and Figure 5-5, respectively. Although greater knee abduction angle has been considered to be a risk factor of

PFPS, subjects with PFPS did not necessarily demonstrate greater knee abduction angle (or smaller knee adduction angle) than asymptomatic controls during running. Rather, greater knee adduction angles were found in the PFPS groups in previous studies (Dierks et al., 2011; Noehren et al., 2012b; Willy et al., 2012). In addition, it has been reported that subjects with PFPS demonstrated increased internal knee abduction impulse and/or moment compared to non-injured controls during running (Stefanyshyn et al., 2006; Willy et al., 2012). Based on the mechanism described in the paragraph above, greater internal knee abduction moment would increase knee adduction angle, and the results of this thesis would be corresponding to these previous findings.

In contrast to the medial collapse mechanism of the knee (Powers, 2003), increasing knee adduction will likely lead to a decrease in the Q-angle. Decreased Q-angle has been shown to result in medial tracking of patella and an increase in contact pressure of the medial facet of the patella (Huberti and Heyes, 1984; Besier et al., 2008). Thus, in the light of these findings, it can be considered that both greater knee abduction and adduction angles have a detrimental effect in the development of PFPS, and greater internal knee abduction impulse and moment are the factors influencing frontal plane knee joint angles.

5-3. Limitations of the experiments and future directions

Limitations and errors associated with the calculation of lower extremity kinematics have been well documented. For the rearfoot segment, external rearfoot markers were placed on the heel counter of the shoe in this study. Reinschmidt et al. (1997) reported that external markers on the shoe are only gross indicators of the skeletal tibiocalcaneal motion. Similarly, it is clear that tracking marker motions relative to the underlying bone might not represent an accurate estimation of segmental kinematics due to soft-tissue artifacts, which is the most significant error in motion analysis (Leardini et al., 2005). In this thesis, we attempted to avoid areas of large muscle mass when attaching the tracking markers. Moreover, all subjects wore the same shoes during running. In Chapter 2, the ICC values for the measured angles were greater than 0.86 for trial-retrial reliability and greater than 0.77 for test-retest reliability, indicating good to excellent reliability. In addition, the mean values of the RMS differences were less than 1.7°. Therefore, although there is a possibility that the joint angles reported in this study over-/underestimate the true joint motion, the measured angles were reliable and the measurement errors were sufficiently smaller than the observed joint angles of interest in the present study. Thus, we expect that this soft-tissue artifact is systematic and does not significantly affect to the findings of this thesis.

Another limitation of this thesis was that the study design of Chapter 3 and Chapter

4 was a cross-sectional in nature and can only provide information regarding associations, and not cause and effect. In addition, only one previous study prospectively investigated the knee joint kinetics of PFPS runners (Stefanyshyn et al., 2006). The results of this thesis revealed that there appears to exist a different mechanism that contradicts to the previously proposed concepts (Chapter 3), and a conflict between the kinematic and kinetic risk factors of PFPS (Chapter 4). More recently, since male subjects with PFPS demonstrated greater peak knee adduction angle and external knee adduction moment compared with females with PFPS and with healthy males, there may exist a different mechanism between genders in the development of PFPS (Willy et al., 2012). Thus, 'a safe range' for the risk factors of running injuries at the knee still remains unclear. However, the current results provide indications how ankle and hip joint kinematics associate with frontal plane knee joint mechanics during running which is generally working on both genders. Therefore, hip external rotation angle may be the key factor to control frontal plane knee joint mechanics. Future studies are needed to see if alteration of hip rotation angle affects knee joint mechanics during running, and to identify the biomechanical 'safety range' of running injuries at the knee.

Some intervention methods, such as hip abductor and external rotator strength training and alteration of running style, have been attempted to alter dynamic ankle, knee and hip joint mechanics during running. Snyder et al. (2009) investigated the effect of 6-weeks of

hip resistance training on lower extremity joint kinematics and kinetics during running, and reported that increased hip muscle strength altered the joint angles and moments at the ankle, knee and hip joints. However, whereas the internal knee abduction moment decreased by 10%, the changes in joint range of motion, which significantly reduced following resistance training, were below 2 degrees. Thus, the effect of resistance training of hip muscles may be small.

Step width has a significant influence on rearfoot kinematics in the frontal plane (Williams and Ziff, 1991; Pohl et al., 2006). These studies reported that peak rearfoot eversion angle was significantly decreased when subjects ran with a wider step width. Also, a wider step width has been speculated to decrease knee joint loading (Stefanyshyn et al., 2006). However, taking into consideration the results of Chapter 2 and Chapter 3, since increasing the step width will result in smaller rearfoot eversion, this alteration of step width may increase knee abduction angle. Therefore, based on the results of this thesis, alteration of hip external rotation angle may be more effective to control knee joint mechanics. Future studies will clarify the alterations of hip rotation angle on ankle and hip joint kinematics and their influence on knee joint mechanics during running.

Frontal plane knee joint mechanics can also be influenced by footwear. The result of this thesis indicated that an increased toe-out angle coupled with hip external rotation, would be beneficial to change knee joint mechanics. However, the duration of the stance phase of

running was within 230 ms when subjects ran at a speed of 3.5 m/s (Chapter 2). In addition, foot does not rotate in the transverse plane during mid-stance (foot rotation angle with respect to the laboratory coordinate system in the transverse plane can be seen in Figure 4-2). Thus, the findings of this thesis suggests that if the functional footwear, which have a function to control toe-in/out angles, work within approximately 10-20% of stance, they could have a potential to reduce the risk of running injuries at the knee.

5-4. Conclusion of the thesis

The general purpose of this thesis was to elucidate possible mechanisms how ankle and hip joint kinematics associate with frontal plane knee joint mechanics during running. To this end, the relationships between ankle and hip joint kinematics on knee joint mechanics were examined. The main findings of this thesis are the following three points: 1) female runners, who are more likely to develop knee injuries, demonstrated smaller rearfoot eversion, greater knee abduction, greater hip adduction and greater hip internal rotation angles, 2) greater knee abduction angle is associated with smaller hip internal rotation, smaller rearfoot eversion and greater hip adduction angles, 3) subject with smaller frontal plane lever arm demonstrated that greater hip external rotation angle resulted in greater toe-out angle, greater AJC-CoP distance, and subsequent smaller knee abduction moment. These findings suggest

that 1) there exists a different mechanism on kinematic risk factor of running injuries at the

knee, and that 2) hip rotation angle may be a key factor on knee joint mechanics. This

information has the potential to understand how ankle and hip joint kinematics are related with

frontal plane knee joint mechanics during running, and to help for the intervention of running

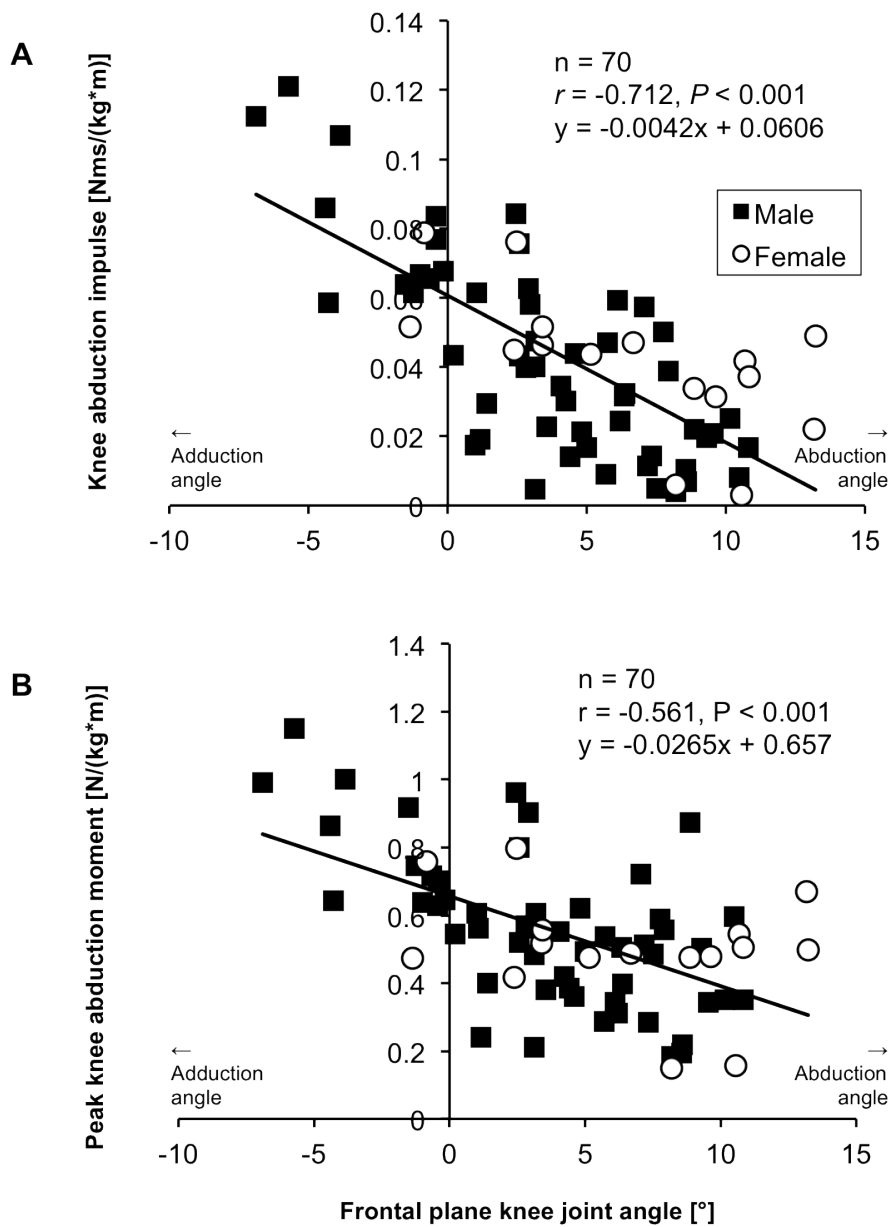style and the development of functional footwear to control knee joint mechanics.

Figure 5-1. Pearson's correlations between frontal plane knee joint angle and (A) knee abduction impulse, and (B) peak knee abduction moment (filled squares are males, open circles are females).
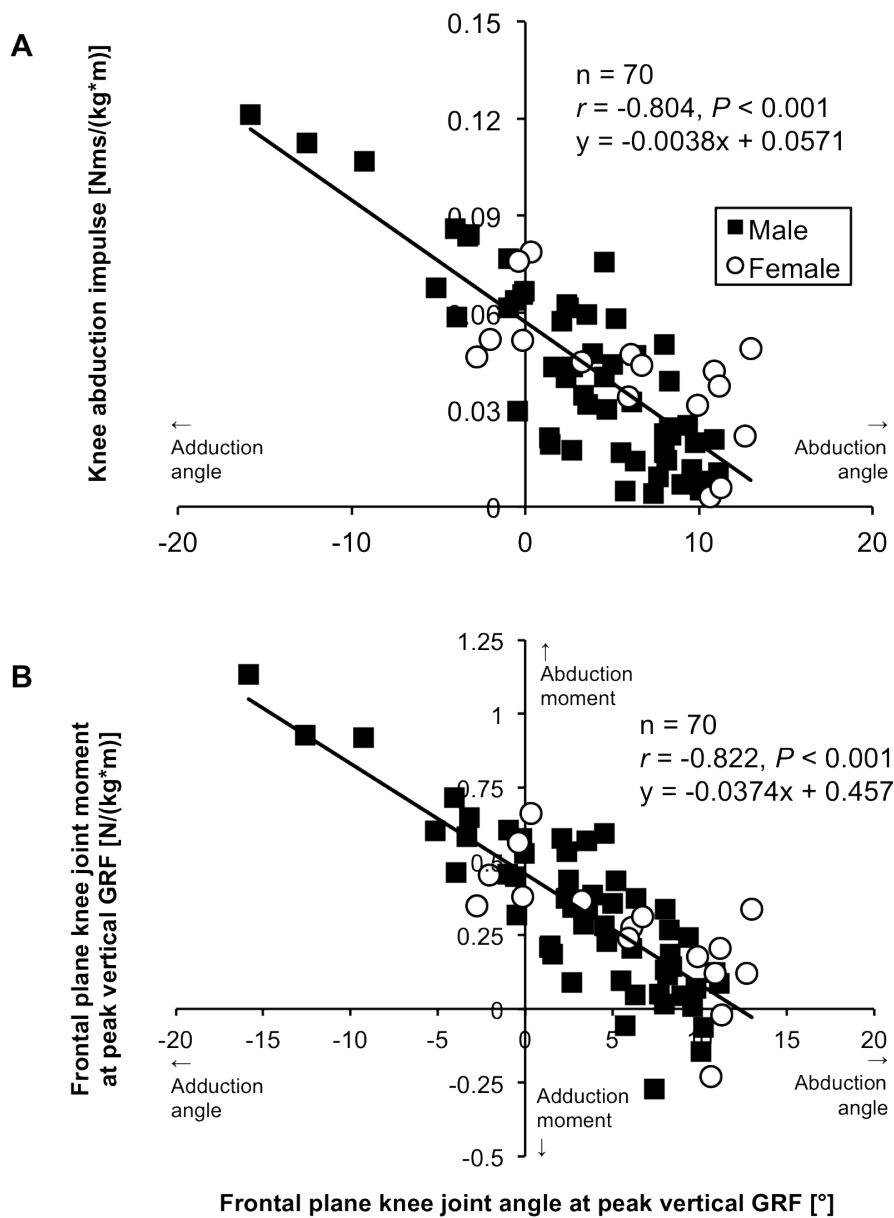
Figure 5-2. Pearson's correlations between frontal plane knee joint angle at peak vertical GRF and (A) knee abduction impulse, and (B) frontal plane knee joint moment at peak vertical GRF (filled squares are males, open circles are females).

A   Greater hip internal rotation

B   Greater hip external rotation

GRF

GRF

Lever arm

Internal knee abduction moment

Center of pressure

AJC-CoP distance

Greater Toe-out
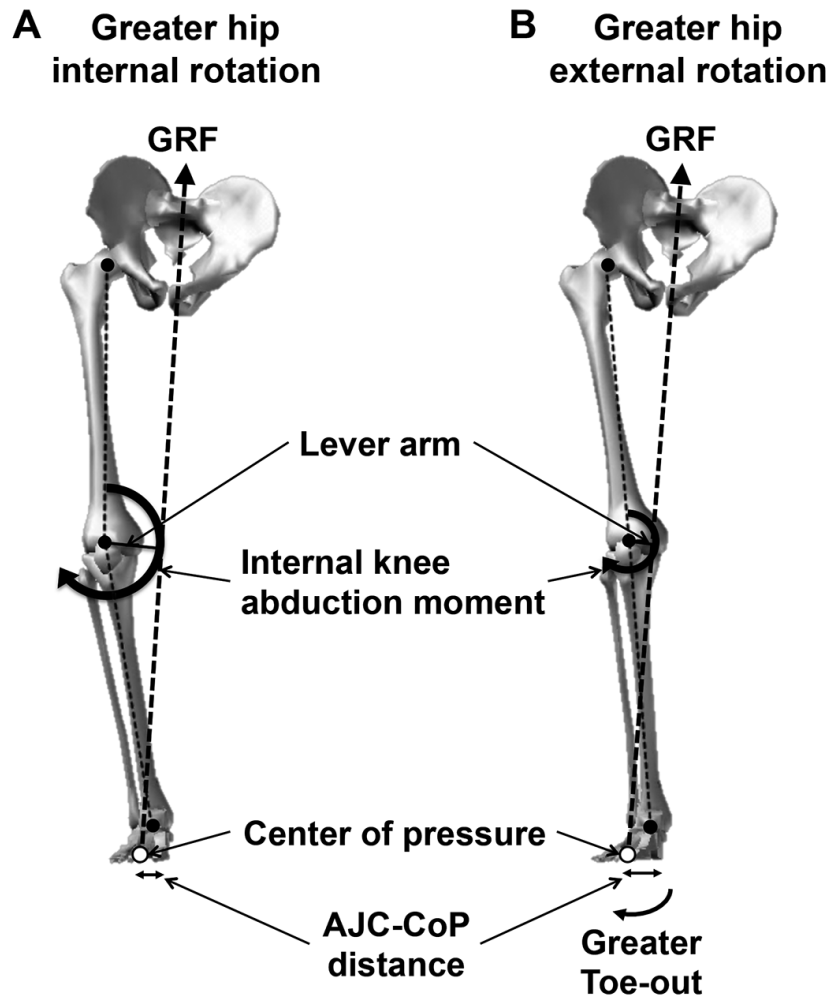
Figure 5-3. Schematic illustrations showing how hip rotation angles affect the knee joint kinematics and kinetics during the stance phase of running.
The subject with greater hip internal rotation angle on the left (A), and the subject with greater hip external rotation angle on the right (B). Greater hip external rotation angle would result in greater toe-out angle, and this movement would decrease the frontal plane lever arm of the knee joint moment resulting in decreased internal knee abduction moment. At the same time, the subject with greater hip external rotation angle would demonstrate greater knee abduction angle and smaller rearfoot eversion angle.
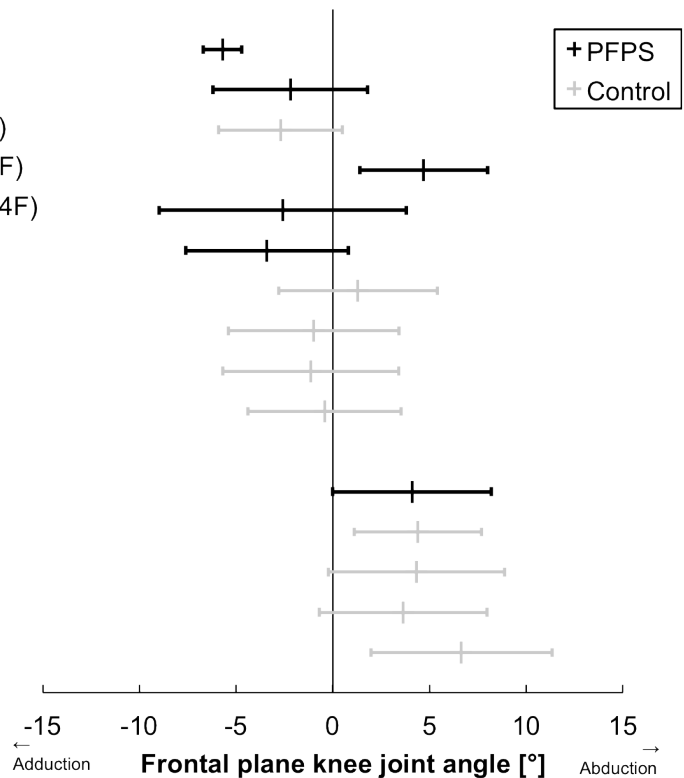
Figure 5-4. Mean values (± SD) of the peak frontal plane knee joint angles during running reported in previous studies compared with this study (black plots are PFPS groups, grey plots are asymptomatic control groups). In the study of Dierks et al. (2011), PFPS subjects were subdivided into 3 groups, PFPS valgus, PFPS hip abd and PFPS others, based on their movement patterns. "M" indicates the number of male subjects, "F" indicates the number of female subjects.
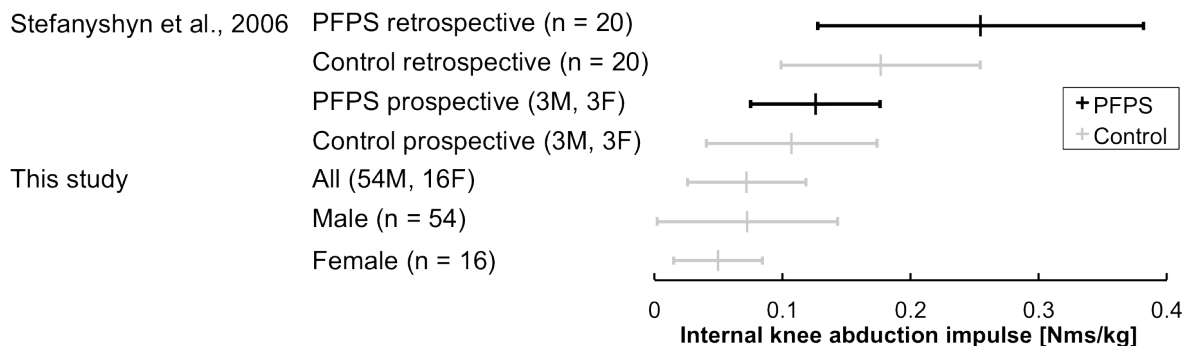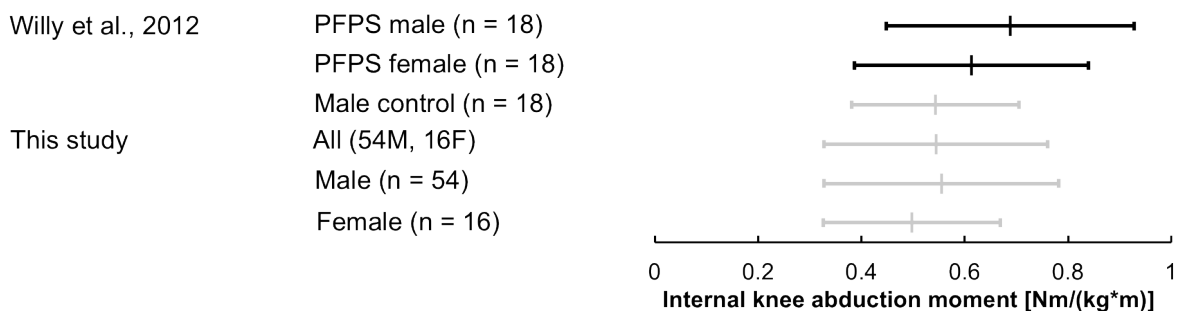
Figure 5-5. Mean values (± SD) of the internal knee abduction moment and impulse during running reported in previous studies compared with this study (black plots are PFPS groups, grey plots are asymptomatic control groups). "M" indicates the number of male subjects, "F" indicates the number of female subjects.

References

Ae M, Tang H, Yokoi T. Estimation of inertia properties of the body segments in Japanese athletes. (in Japanese) *Biomechanisms 11*, Society of Biomechanisms, University of Tokyo Press, Tokyo, pp. 23-33, 1992.

Akimoto H. Running injuries in ultra-marathon runners. (in Japanese) *Japanese Journal of Clinical Sports Medicine,* 15: 26-32, 2007.

Andrews JG. Euler's and Lagrange's equations for linked rigid-body models of three-dimensional human motion. In Allard P, Stokes IAF, Blanchi JP, editors. *Three-dimensional analysis of human movement*, Human Kinetics, Champaign (IL), pp. 145-175, 1995.

Andrews M, Noyes FR, Hewett TE, Andriacchi TP. Lower limb alignment and foot angle are related to stance phase knee adduction in normal subjects: A critical analysis of the reliability of gait analysis data. *Journal of Orthopaedic Research*, 14: 289-295, 1996.

Areblad M, Nigg BM, Ekstrand J, Olsson KO, Ekström H. Three-dimensional measurement of rearfoot motion during running. *Journal of Biomechanics,* 23: 933-940, 1990.

Bell AL, Brand RA, Pedersen DR. Prediction of hip joint centre location from external

landmarks. *Human Movement Science,* 6:3-16, 1989.

Besier TF, Gold GE, Delp SL, Fredericson M, Beaupré GS. The influence of femoral internal and external rotation on cartilage stresses within the patellofemoral joint. *Journal of Orthopaedic Research*, 26:1627-1635, 2008.

Carry PM, Kanai S, Miller NH, Polousky JD. Adolescent patellofemoral pain: a review of evidence for the role of lower extremity biomechanics and core instability. *Orthopedics,* 33:498-507, 2010.

Chiari L, Della Croce U, Leardini A, Cappozzo A. Human movement analysis using stereophotogrammetry. Part 2: instrumental errors. *Gait & Posture,* 21:197-211, 2005.

Chumanov ES, Wall-Scheffler C, Heiderscheit BC. Gender differences in walking and running on level and inclined surfaces. *Clinical Biomechanics,* 23: 1260-1268, 2008.

Chuter VH, Janse de Jonge XA. Proximal and distal contributions to lower extremity injury: a review of the literature. *Gait & Posture*, 36:7-15, 2012.

Cichanowski HR, Schmitt JS, Johnson RJ, Niemuth PE. Hip strength in collegiate female athletes with patellofemoral pain. *Medicine & Science in Sports & Exercise,* 39:1227-1232, 2007.

Clement DB, Taunton JE, Smart GW, McNicol KL. A survey of overuse running injuries. *The Physician and Sportsmedicine,* 9:47-58, 1981.

Cohen, J. *Statistical power analysis for the behavioral sciences*. 3rd ed. Hillsdale, NJ: Lawrence Erlbaum Associates, 1988.

Cole GK, Nigg BM Ronsky JL Yeadon MR. Application of the joint coordinate system to three-dimensional joint attitude and movement representation: a standardization proposal. *Journal of Biomechanical Engineering,* 115:344-349, 1993.

Davis IS, Powers CM. Patellofemoral pain syndrome: proximal, distal, and local factors, an international retreat. *Journal of Orthopaedic & Sports Physical Therapy,* 40:A1-16, 2010.

DeLeo AT, Dierks TA, Ferber R, Davis IS. Lower extremity joint coupling during running: a current update. *Clinical Biomechanics,* 19:983-991, 2004.

Dierks TA, Manal KT, Hamill J, Davis IS. Proximal and distal influences on hip and knee kinematics in runners with patellofemoral pain during a prolonged run. *Journal of Orthopaedic & Sports Physical Therapy,* 38:448-456, 2008.

Dierks TA, Manal KT, Hamill J, Davis I. Lower extremity kinematics in runners with patellofemoral pain during a prolonged run. *Medicine & Science in Sports & Exercise,* 43:693-700, 2011.

Duffey MJ, Martin DF, Cannon DW, Craven T, Messier SP. Etiologic factors associated with anterior knee pain in distance runners. *Medicine & Science in Sports & Exercise,*

32:1825-1832, 2000.

Dugan SA, Bhat, KP. Biomechanics and analysis of running gait. *Physical Medicine and Rehabilitation Clinics of North America*, 16:603-621, 2005.

Ferber R, Davis IM, Williams III DS. Gender differences in lower extremity mechanics during running. *Clinical Biomechanics,* 18:350-357, 2003.

Ferber R, Hreljac A, Kendall KD. Suspected mechanisms in the cause of overuse running injuries: a clinical review. *Sports Health: A Multidisciplinary Approach,* 1:242-246, 2009.

Ferber R, Noehren B, Hamill J, Davis IS. Competitive female runners with a history of iliotibial band syndrome demonstrate atypical hip and knee kinematics. *Journal of Orthopaedic & Sports Physical Therapy,* 40:52-58, 2010.

Fisher DS, Dyrby CO, Mündermann A, Morag E, Andriacchi TP. In healthy subjects without knee osteoarthritis, the peak knee adduction moment influences the acute effect of shoe interventions designed to reduce medial compartment knee load. *Journal of Orthopaedic Research,* 25:540-546, 2007.

Fulkerson JP. Diagnosis and treatment of patients with patellofemoral pain. *American Journal of Sports Medicine,* 30:447-456, 2002.

Grood ES, Suntay WJ. A joint coordinate system for the clinical description of

three-dimensional motions: application to the knee. *Journal of Biomechanical Engineering,* 105:136-144, 1983.

Heiderscheit BC, Hamill J, Caldwell GE. Influence of Q-angle on lower-extremity running kinematics. *Journal of Orthopaedic & Sports Physical Therapy,* 30:271-278, 2000.

Hinman RS, Bowles KA, Payne C, Bennell KL. Effect of length on laterally-wedged insoles in knee osteoarthritis. *Arthritis & Rheumatism,* 59:144-147, 2008.

Hintermann B, Nigg BM. Pronation in runners. Implications for injuries. *Sports Medicine,* 26:169-176, 1998.

Horton MG, Hall TL. Quadriceps femoris muscle angle: normal values and relationships with gender and selected skeletal measures. *Journal of Orthopaedic & Sports Physical Therapy,* 69:897-901, 1989.

Hreljac A, Marshall RN, Hume PA. Evaluation of lower extremity overuse injury potential in runners. *Medicine & Science in Sports & Exercise,* 32:1635-1641, 2000.

Huberti HH, Hayes WC. Patellofemoral contact pressures-the influence of Q-angle and tendofemoral contact. *The Journal of Bone & Joint Surgery,* 66:715-724, 1984.

Hunt MA, Birmingham TB, Giffin JR Jenkyn TR. Associations among knee adduction moment, frontal plane ground reaction force, and lever arm during walking in patients with knee osteoarthritis. *Journal of Biomechanics,* 39:2213-2220, 2006.

Ireland ML, Willson JD, Ballantyne BT, Davis IM. Hip strength in females with and without patellofemoral pain. *Journal of Orthopaedic & Sports Physical Therapy,* 33:671-676, 2003.

James SL, Bates BT, Osternig LR. Injuries to runners. *American Journal of Sports Medicine,* 6:40-50, 1978.

Jenkyn TR, Hunt MA, Jones IC, Giffin JR, Birmingham TB. Toe-out gait in patients with knee osteoarthritis partially transforms external knee adduction moment into flexion moment during early stance phase of gait: a tri-planar kinetic mechanism. *Journal of Biomechanics,* 41:276-283, 2008.

Jenkyn TR, Erhart JC, Andriacchi TP. An analysis of the mechanisms for reducing the knee adduction moment during walking using a variable stiffness shoe in subjects with knee osteoarthritis. *Journal of Biomechanics,* 44:1271-1276, 2011.

Kosmahl EM, Kosmahl HE. Painful plantar heel, plantar fasciitis, and calcaneal spur: etiology and treatment. *Journal of Orthopaedic & Sports Physical Therapy,* 9:17-24, 1987.

Kvist M. Achilles tendon injuries in athletes. *Sports Medicine,* 18:173-201, 1994.

Leardini A, Chiari L, Della Croce U, Cappozzo A. Human movement analysis using stereophotogrammetry. Part 3. Soft tissue artifact assessment and compensation. *Gait &*

Posture, 21:212-225, 2005.

Leetun DT, Ireland ML, Willson BT, Ballantyne BT, Davis IM. Core stability measures as

risk factors for lower extremity injury in athletes. *Medicine & Science in Sports &*

Exercise. 36:926-934, 2004.

Lun V, Meeuwisse WH, Stergiou P, Stefanyshyn DJ. Relation between running injury and

static lower limb alignment in recreational runners. *British Journal of Sports Medicine,*

38:576-580, 2004.

Malinzak RA, Colby SM, Kirkendall DT, Yu B, Garrett WE. A comparison of knee joint

motion patterns between men and women in selected athletic tasks. *Clinical*

Biomechanics, 16:438-445, 2001.

McClay I. The evolution of the study of the mechanics of running: relationship to injury.

Journal of the American Podiatric Medical Association, 90:133-148, 2000.

McClay I, Manal K. Coupling parameters in runners with normal and excessive pronation.

Journal of Applied Biomechanics. 13:109-124, 1997.

McClay I, Manal K. A comparison of three-dimensional lower extremity kinematics during

running between excessive pronators and normal. *Clinical Biomechanics,* 13:195-203,

1998.

McClay I, Manal K. Three-dimensional kinetic analysis of running: significance of secondary

planes of motion. *Medicine &Science in Sports & Exercise,* 31:1629-1637, 1999.

Messier SP, Pittala KA. Etiologic factors associated with selected running injuries. *Medicine & Science in Sports & Exercise,* 20:501-505, 1988.

Messier SP, Davis SE, Curl WW, Lowery RB, Pack RJ. Etiologic factors associated with patellofemoral pain in runners. *Medicine & Science in Sports & Exercise,* 23:1008-1015, 1991.

Messier SP, Edwards DG, Martin DF, Lowery RB, Cannon DW, James MK, Curl WW, Read HM Jr, Hunter DM. Etiology of iliotibial band friction syndrome in distance runners. *Medicine & Science in Sports & Exercise,* 27:951-960, 1995.

Mizuno Y, Kumagai M, Mattessich SM, Elias JJ, Ramrattan N, Cosgarea AJ, Chao EY. Q-angle influences tibiofemoral and patellofemoral kinematics. *Journal of Orthopaedic Research,* 19:834-840, 2000.

Niemuth PE, Johnson RJ, Myers MJ, Thieman TJ. Hip muscle weakness and overuse injuries in recreational runners. *Clinical Journal of Sports Medicine,* 15:14-21, 2005.

Noakes TD. *Lore of Running.* 4th ed. Champaign (IL): Human Kinetics; 2004.

Noehren B, Davis I, Hamill J. Prospective study of the biomechanical factors associated with iliotibial band syndrome. *Clinical Biomechanics,* 22:951-956, 2007.

Noehren B, Pohl MB, Sanchez Z, Cunningham T, Lattermann C. Proximal and distal

kinematics in female runners with patellofemoral pain. *Clinical Biomechanics,* 27:366-371, 2012a.

Noehren B, Sanchez Z, Cunningham T, McKeon PO. The effect of pain on hip and knee kinematics during running in females with chronic patellofemoral pain. *Gait & Posture,* 36:596-599. 2012b.

Noguchi J. Running injuries to joggers. (in Japanese) *Japanese Journal of Clinical Sports Medicine,* 8:92-94, 2000

Park SK, Stefanyshyn DJ, Ramage B, Hart DA, Ronsky JL. Alterations in knee joint laxity during the menstrual cycle in healthy women leads to increases in joint loads during selected athletic movements. *American Journal of Sports Medicine,* 37:1169-1177, 2009.

Park SK, Stefanyshyn DJ. Greater Q angle may not be a risk factor of patellofemoral pain syndrome. *Clinical Biomechanics,* 26:392-396, 2011.

Pohl MB, Messenger N, Buckley JG. Changes in foot and lower limb coupling due to systematic variations in step width. *Clinical Biomechanics,* 21:175-183, 2006.

Pohl MB, Mullineaux DR, Milner CE, Hamill J, Davis IS. Biomechanical predictors of retrospective tibial stress fractures in runners. *Journal of Biomechanics,* 41:1160-1165, 2008.

Powers CM. The influence of altered lower-extremity kinematics on patellofemoral joint

dysfunction: a theoretical perspective. *Journal of Orthopaedic & Sports Physical Therapy,* 33:639-646, 2003.

Powers CM, Bolgla LA, Callaghan MJ, Collins N, Sheehan FT. Patellofemoral pain: proximal, distal, and local factors, 2nd International Research Retreat. *Journal of Orthopaedic & Sports Physical Therapy,* 42:A1-54, 2012.

Quatman CE, Hewett TE. The anterior cruciate ligament injury controversy: is "valgus collapse" a sex-specific mechanism? *British Journal of Sports Medicine*, 43:328-335, 2009.

Reiman MP, Bolgla LA, Lorenz D. Hip functions influence on knee dysfunction: a proximal link to a distal problem. *Journal of Sport Rehabilitation*, 18:33-46, 2009.

Reinschmidt C, van den Bogert AJ, Murphy N, Lundberg A, Nigg BM. Tibiocalcaneal motion during running, measured with external and bone markers. *Clinical Biomechanics,* 12:8-16, 1997.

Robertson GE, Caldwell G, Hamill J, Kamen G, Whittlesey S. *Research Methods in Biomechanics.* Human Kinetics, Champaign (IL), 2004.

Russell EM, Hamill J. Lateral wedges decrease biomechanical risk factors for knee osteoarthritis in obese women. *Journal of Biomechanics,* 44:2286-2291, 2011.

Sasakawa Sports Foundation. The 2010 SSF national sports-life survey. (in Japanese)

Sasakawa Sports Foundation, Tokyo, Japan, 2010.

Schache AG, Blanch P, Rath D, Wrigley T, Bennell K. Differences between the sexes in the three-dimensional angular rotations of the lumbo-pelvic-hip complex during treadmill running. *Journal of Sports Sciences,* 21:105-118, 2003.

Schache, AG, Baker, R. On the expression of joint moments during gait. *Gait & Posture,* 25: 440-452, 2007.

Snyder, KR, Earl, JE, O'Connor, KM, Ebersole, KT. Resistance training is accompanied by increases in hip strength and changes in lower extremity biomechanics during running. *Clinical Biomechanics,* 24: 26-34, 2009.

Söderkvist I, Wedin PA. Determining the movements of the skeleton using well-configured markers. *Journal of Biomechanics,* 26:1473-1477, 1993.

Souza RB, Powers CM. Predictors of hip internal rotation during running: an evaluation of hip strength and femoral structure in women with and without patellofemoral pain. *American Journal of Sports Medicine,* 37:579-587, 2009.

Stacoff A, Nigg BM, Reinschmidt C, van den Bogert AJ, Lundberg A. Tibiocalcaneal kinematics of barefoot versus shod running. *Journal of Biomechanics,* 33:1387-1395, 2000.

Stefanyshyn DJ, Stergiou P, Lun VM, Meeuwisse WH, Worobets JT. Knee angular

impulse as a predictor of patellofemoral pain in runners. *American Journal of Sports Medicine,* 34:1844-1851, 2006.

Taunton JE, Ryan MB, Clement DB, McKenzie DC, Lloyd-Smith DR, Zumbo BD. A retrospective case-control analysis of 2002 running injuries. *British Journal of Sports Medicine,* 36:95-101, 2002.

Tiberio D. The effect of excessive subtalar joint pronation on patellofemoral mechanics: a theoretical model. *Journal of Orthopaedic & Sports Physical Therapy,* 9:160-165, 1987.

Vaughan CL, Davis BL, O'Connor JC. *Dynamics of human gait.* Champaign(IL): Human Kinetics; 1992.

van Gent RN, Siem D, van Middelkoop M, van Os AG, Bierma-Zeinstra, SMA, Koes BW. Incidence and determinants of lower extremity running injuries in long distance runners: a systematic review. *British Journal of Sports Medicine*, 41:469-480, 2007.

van Mechelen W. Running injuries: a review of the epidemiological literature. *Sports Medicine,* 14:320-335, 1992.

Wen DY, Puffer JC, Schmalzried TP. Lower extremity alignment and risk of overuse injuries in runners. *Medicine & Science in Sports & Exercise,* 29:1291-1298, 1997.

Wen DY, Puffer JC, Schmalzried TP. Injuries in runners: a prospective study of alignment. *Clinical Journal of Sports Medicine,* 8:187-194, 1998.

Willems TM, Witvrouw EC, De Cook A, De Clercq D. Gait-related risk factors for exercise-related lower-leg pain during shod running. *Medicine & Science in Sports & Exercise,* 30:330-339, 2007.

Williams KR, Ziff JL. Changes in distance running mechanics due to systematic variations in running style. *Journal of Applied Biomechanics,* 7:76-90, 1991.

Williams DS, Davis IM, Baitch SP. Effect of inverted orthoses on lower-extremity mechanics in runners. *Medicine & Science in Sports & Exercise,* 35:2060-2068, 2003.

Willson JD, Davis IS. Lower extremity mechanics of females with and without patellofemoral pain across activities with progressively greater task demands. *Clinical Biomechanics,* 23:203-211, 2008a.

Willson JD, Davis IS. Utility of the frontal plane projection angle in females with patellofemoral pain. *Journal of Orthopaedic & Sports Physical Therapy,* 38:606-615, 2008b.

Willson JD, Davis IS. Lower extremity strength and mechanics during jumping in women with patellofemoral pain. *Journal of Sport Rehabilitation,* 18:76-90, 2009.

Willy RW, Manal KT, Witvrouw EE, Davis IS. Are Mechanics Different between Male and Female Runners with Patellofemoral Pain? *Medicine & Science in Sports & Exercise,* 44:2165-2171, 2012.

Winter DA. *Biomechanics and motor control of human movement.* 3rd ed. Hoboken (NJ):

John Wiley and Sons; 2005.

Wu G, Siegler S, Allard P, Kirtley C, Leardini A, Rosenbaum D, Whittle M, D'Lima DD, Cristofolini L, Witte H, Schmid O, Stokes I. ISB recommendation on definitions of

joint coordinate system of various joints for the reporting of human joint motion-part I:

ankle, hip, and spine. *Journal of Biomechanics,* 35:543-548, 2002.

Acknowledgement

Finally, I would like to show appreciation for my family and friends. They have given me moral support during my research activities.