A CLINICAL AND BIOMECHANICAL PROFILE OF FEMALE ATHLETES WITH AND WITHOUT PATELLOFEMORAL PAIN

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Dedication

This dissertation is dedicated to my parents who always planned that I would get a college degree; little did they know I would end up with four.

To my family, Diane, Sophia, Olivia and Jackson; you are my motivation for what I do each day and my true reason for being.

Abstract

Female athletes may exhibit unique risks and orthopedic presentations when engaging in physical activity. Gender differences have been noted in the incidences of particular knee injuries, with female athletes/runners reported to be two times more likely to present with patellofemoral disorders as their male counterparts. Unfortunately, the etiology of patellofemoral pain (PFP) remains elusive and multifactorial. The purpose of this study was to examine the clinical and biomechanical profiles of female athletes with and without PFP during walking and running to determine if there is a set of variables that differentiates the groups. Forty-three subjects with moderate PFP and forty-five control subjects with no history of PFP participated in one 60 minute testing session. Each participant completed a series of questionnaires and was assessed for height, weight and activity level. A Modified Thomas test, Modified Ober test and straight leg raise assessment were performed bilaterally looking for differences between the right and left limbs. Pelvis and hip range of motion, maximum vertical ground reaction force, contact time, and center of pressure variability (COPx) were measured during 30 second treadmill walk and run trials utilizing a 9-camera motion analysis system and a Pedar insole system. The symmetry index (SI) (Robinson et al., 1987) was used to characterize asymmetry of kinematic and kinetic variables of both groups. Two clinical indices [Tightness Index (TI) and Pelvic Symmetry Index (PSI)] were established based on results of the Modified Thomas and Ober tests. Based on logistic regression results, we found that the TI and PSI clearly differentiated the two groups with the PFP group displaying significantly greater numbers of positive signs within both indices ($p<.001$). Biomechanical variables did not clearly define the groups, however, there was some evidence that measurement of the mediolateral variability of the COPx/gait line might indicate lower extremity kinetic chain instability as seen at the foot and ankle during activity (p=.07 during walking; p=.013 during running). The results of this study suggest that there is a significant association between the results of the TI and PSI and PFP. Clinically, the tests are quick and easy to perform and may be helpful in distinguishing those at risk for PFP.

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CHAPTER I

Introduction

Patellofemoral pain (PFP) is a common problem in active and athletic individuals and one of the most common musculoskeletal disorders affecting both young and adult populations. Incidence rates of 25% to 40% have been reported by most orthopedic and sports medicine centers (Bizzini, Childs, Piva, & Delitto, 2003; Crossley, Bennell, Green, Cowan, & McConnell, 2002; Devereaux & Lachmann, 1984; Ireland, Willson, Ballantyne, & Davis, 2003). Information regarding gender differences is clear; PFP and dysfunction are seen more commonly in women in general (Csintalan, Schulz, Woo, McMahon, & Lee, 2002; Fulkerson, 2002; Ireland et al., 2003; Prather & Hunt, 2005) and more specifically in women athletes as compared with male athletes (Baker & Juhn, 2000; DeHaven & Lintner, 1986; Devereaux & Lachmann, 1984; Taunton et al., 2002). Women runners are reported to be twice as likely to present with PFP as compared to their male counterparts (Ferber, Davis, & Williams, 2003; Taunton et al., 2002).

There is no consensus in the medical literature concerning definition, etiology, and diagnosis of PFP. Many terms have been used to describe PFP including anterior knee pain, runner’s knee, patellofemoral arthralgia, extensor mechanism dysfunction, medial facet syndrome, lateral facet syndrome, lateral compression syndrome, and patellar malalignment syndrome (Crossley et al., 2002; Csintalan et al., 2002). Crossley and associates (2002) broadly define “patellofemoral pain to encompass all anterior or retropatellar knee pain in the absence of other identifiable pathologic conditions” (p.858). This patellofemoral pain is associated with an insidious onset of diffuse or local knee pain made worse with activities such as prolonged sitting, going up and down stairs, kneeling, squatting, running and jumping (Arroll, Ellis-Pegler, Edwards, & Sutcliffe, 1997; Heintjes et al., 2003; Witvrouw, Lysens, Bellemans, Cambier, & Vanderstraeten, 2000).

Despite its high prevalence, the etiology of PFP remains elusive and treatment approaches remain controversial. Most authors cite a combination of intrinsic and extrinsic variables as being etiological factors to PFP. Included in these factors are quadriceps weakness or insufficiency, insufficient hamstring activation with associated hip flexor tone, patellar malalignment and/or abnormal patellar tracking, increased PF joint stress, poor flexibility, excessive pronation, sudden increases in mileage or training intensity, and poor equipment or training surfaces (Brechter & Powers, 2002a; Hruska, 1998; Messier, Davis, Curl, Lowery, & Pack, 1991; Powers, 2003; Timm, 1998; Witvrouw et al., 2000). Historically, conservative therapy has focused on vastus medialis oblique (VMO) and quadriceps strengthening, patellar taping and bracing, stretching and open and closed kinetic chain exercises. Outcome measures reported have included decrements in pain, improved quadriceps activation as measured by electromyographic activity, and improved outcomes during functional stepping and squatting tasks (Bizzini et al., 2003; Cowan, Hodges, Bennell, & Crossley, 2002; Crossley, Bennell, Green, & McConnell, 2001; Crossley et al., 2002; Powers, 1998). Unfortunately, 70% of conservatively treated PFP patients are symptomatic 12 months following these interventions, indicating poor efficacy in these traditional treatment approaches (Devereaux & Lachmann, 1984).

There are multiple factors in addition to the patella and quadriceps muscles that play key roles in the malalignment and pain presentations of PFP. Numerous authors have recognized the vital link between the proximal and distal segmental interactions of the lower extremity as it relates to the lumbo-pelvic-hip complex or pelvifemoral complex. Abnormal motion(s) of the pelvis, femur and tibia in all three planes of motion are believed to have significant effects on patellofemoral joint mechanics and therefore PFP (Grelsamer & Klein, 1998; Hruska, 1998; Powers, 2003). Recent orthopedic physical therapy investigations look at how the effects of hip strength and flexibility contribute to pelvic control and leg alignment in individuals with and without PFP (Ireland et al., 2003; Mascal, Landel, & Powers, 2003; Piva, Goodnite, & Childs, 2005; Tyler, Nicholas, Mullaney, & McHugh, 2006; Willson, Ireland, & Davis, 2006). While there is more interest in the clinical presentation of the pelvis, hip and femur as they relate to knee pain, there is little research available exploring the biomechanical impact that lack of pelvifemoral control has on lower extremity function. Numerous studies, both static and dynamic, have incorporated kinematic and/or kinetic data from healthy subjects and small PFP patient samples to determine patellar alignment, quadriceps forces and joint specific (e.g. patellofemoral joint reaction forces) measures during activities (Brechter & Powers, 2002a; Brechter & Powers, 2002; MacIntyre, Hill, Fellows, Ellis, & Wilson, 2006; Powers, 2000a; Powers, 2000b). More recently, a few studies have begun to incorporate similar analyses into more functional movement patterns of the entire pelvifemoral complex (McCrory, Quick, Shapiro, Ballantyne, & McClay Davis, 2004; Powers, Heino, Rao, & Perry, 1999; Powers, Chen, Reischl, & Perry, 2002; Powers, 2003).

While no universal definition for PFP exists, most authors do agree that PFP is multifactorial in nature and is considered an overuse or overload syndrome of the knee and lower extremity secondary to biomechanical, structural and training factors (Baker & Juhn, 2000; Fulkerson, 2002; Hreljac, 2005; McClay, 2000). Biomechanical evaluations of walking and running have the potential to identify risk factors or patterns of dysfunction and contribute to the understanding and prevention of overuse musculoskeletal injuries (Schache, Bennell, Blanch, & Wrigley, 1999). Most of the biomechanical research has focused on the lower leg complex, particularly movement patterns of the knee, ankle and subtalar joint and their coordinated movement patterns during running. Significantly less research is available on the integrated biomechanical function of the lumbo-pelvic-hip complex or pelvifemoral complex during walking and running (Schache et al., 1999; Schache et al., 2001; Schache et al., 2002). In addition, gender differences in pelvifemoral motion during walking, running, and functional tasks have been minimally investigated (Decker, Torry, Wyland, Sterett, & Steadman, 2003; Ferber et al., 2003; Kerrigan, Todd, & Della Croce, 1998; Schache, Blanch, Rath, Wrigley, & Bennell, 2003; Smith, Lelas, & Kerrigan, 2002).

Statement of the Problem

Female athletes may present with unique risks and orthopedic presentations when engaging in physical activity. It is evident that anatomic differences which are unique to the female athlete contribute to biomechanical differences affecting structure, alignment and function (McClay, 2000; Prather & Hunt, 2005) and frequently lead to incidences of pelvifemoral dysfunction and PFP. Unfortunately, traditional clinical PFP assessment techniques are inconsistent and choice of special tests and the associated outcomes vary significantly between examining therapists. In addition, no single or multiple biomechanical factors have been consistently shown to predict the presence or outcome of PFP syndrome (Baker & Juhn, 2000; Heintjes et al., 2003). Currently, there is minimal research exploring the biomechanical impact that pelvifemoral instability in a PFP group has on lower extremity kinematics and kinetics and what their associated relationships are to clinical assessment tests. It has been suggested that multifactorial models should be generated to help identify which mechanical factors are associated with specific injuries to more optimally predict who may be at risk for activity-related injury (McClay, 2000).

The objective of this study was to investigate the following research questions:

1. Do standard pelvis, hip and knee clinical assessment tests clearly differentiate female PFP subjects from healthy, female control subjects?

2. Are there noticeable differences or asymmetries between the right and left legs of PFP subjects as compared to the control subjects?

3. Do individuals with pelvifemoral dysfunction/PFP demonstrate pelvic and femoral kinematic differences during walking and running as compared to healthy, female control subjects?

4. What force distributions (kinetics) are exhibited on the lower extremities of female subjects with pelvifemoral dysfunction and PFP during walking and running as compared to healthy, female control subjects?

Based upon these four questions, the specific aims of this study are listed below. In order to achieve the specific aims, the associated hypotheses were tested.

Aim 1. Establish a clinical profile of symptomatic female PFP subjects.

- To assess the results of clinical assessment tests as suggested by the Postural Restoration Institute® to establish pelvic/femur alignment in female athletes with PFP versus asymptomatic athletes and to compare them to measured kinematic and kinetic data.
    - Hypothesis 1: PFP subjects will display a greater frequency of positive results on both the Modified Thomas/Extension Drop Test and the Ober/Adduction Drop Test and will present with greater asymmetry on a bilateral straight leg raise (SLR) assessment, all measures of pelvic asymmetry.

Aim 2: Establish a biomechanical profile of symptomatic female PFP subjects.

- To assess 3-D kinematics of the pelvis, hips and femurs in female athletes with PFP versus asymptomatic female athletes in a walking and running condition.
    - Hypothesis 2: Ground contact time distribution patterns will be different between the PFP and control groups.
    - Hypothesis 3: Anterior pelvic tilt and hip extension range of motion (ROM) measurements will be different between the PFP and control groups.
- To assess a kinetic analysis of female athletes with PFP versus asymptomatic female athletes in a walking and running condition.
    - Hypothesis 4: Significant kinetic differences will be present between PFP subjects and controls as measured by maximum

ground reaction force distributions and center of pressure patterns (between the right and left lower extremities).

Aim 3. Determine if there is a set of biomechanical variables that discriminate between groups of individuals with and without PFP that could guide future research regarding evaluation and diagnosis of these patients, as well as establish improved outcomes through more effective rehabilitation programs.

Hypothesis 5: There will be a distinct group of biomechanical variables that will differentiate PFP subjects from control subjects during both the walking and running conditions.

CHAPTER II

Review of Literature

The causes of chronic overuse musculoskeletal injuries and specifically, pelvifemoral and/or PFP injuries are varied and diverse. Clinical gait analysis has great potential for evaluating and treating musculoskeletal injuries and syndromes; however, much of the information on running and walking mechanics is scattered throughout the scientific literature rather than the clinical literature and much of it does not consider the issues of clinical application (Andriacchi, 1990; Cavanagh, 1987). Because some adaptations to orthopedic dysfunction appear only during locomotion, a functional evaluation must be used to identify and quantify abnormalities in function and/or motion (Andriacchi, 1990). Often, a biomechanical kinetic and kinematic analysis of these adaptations does more than conventional clinical measures alone to reveal the nature of the underlying dysfunction or abnormality of the movement pattern (e.g. force or pressure distributions of the foot and lower extremity during weight bearing). Clinical use of gait analysis, whether it is for assessing neurological, orthopedic, or sports related injuries, requires an understanding of the functional adaptations associated with the dysynchronies of movement. Although some gait abnormalities are a direct result of a mechanical change brought on by the injury and are easy to objectively evaluate, in many cases they represent an adaptation or compensation pattern of the pathologic condition which may not be immediately evident (Andriacchi, 1990). Understanding the nature of the underlying abnormality or dysfunction is the goal of evaluation and treatment of musculoskeletal disorders. However, before we can be effective in assessing pathology or dysfunction of movement, it is critical to have a fundamental understanding of normal walking and running gait. Therefore, the first part of this literature review will focus on a review of the biomechanics of walking and running gait as they relate to the lower extremity kinetic chain, more specifically, the pelvis, hips and knees.

While proper walking and running biomechanics involves synchronous movements of all the components of both the upper and lower kinetic chains of the body, much of the earlier running related biomechanics research focused on kinetic and kinematic analyses related to the foot and lower leg only (Cavanagh, 1987; Dugan & Bhat, 2005; Hreljac, Marshall & Hume, 2000). More recently, researchers have begun to look up the kinetic chain in their search for factors related to overuse running injuries. The second part of the literature review will look more specifically at the coordinated movement patterns of the lumbo-pelvic-hip complex as it relates to walking and running. The review will also focus on specific studies of kinematic and kinetic presentations related to gender differences and the patellofemoral joint. The final section will review research on asymmetries in normal human gait, as well as the reliability and limitations of systems used to measure kinetic and kinematic variables of symmetry or asymmetry during gait and functional task assessments.

Biomechanics of Walking and Running

Gait Cycle

Normal gait is very complex, yet most individuals perform the activity in what appears to be a very similar, synchronous and symmetrical manner. Human locomotion can be divided into walking, running and sprinting. These three conditions represent a continuum of movement ranging from slow walking to sprinting at top speeds. Running is a natural extension of walking. While running is typically defined or described by the velocity achieved by the runner, the actual distinction between walking and running is made according to the percentage of stance time, or ground contact, in the gait cycle (Gage, 1990; Novacheck, 1995; Ounpuu, 1990). When describing a subject's gait pattern, it is conventional to do so in terms of the gait cycle (Figure 1). The gait cycle is the period from the initial contact of one foot to the following initial contact of the same foot. It is divided into the stance and swing phases with stance usually representing 60% and swing 40% of the gait cycle in normal walking (Gage, 1990; Mann & Hagy, 1980; Novacheck, 1995; Ounpuu, 1990). The stance phase (Figure 1A) begins with the initial contact of one foot and is divided into subphases: loading response, midstance, terminal stance, and preswing (Ounpuu, 1994). Loading response is a period of deceleration when the shock of impact is absorbed (Gage, 1990). This is followed by a period of single leg stance occupying 40% of the cycle, during which time the opposite limb is going through its swing phase (Gage, 1990). The swing phase, which begins with toe-off of the contralateral foot and ends with initial contact, is divided into three subphases: initial swing, midswing, and terminal swing (Novacheck, 1995; Ounpuu, 1994). During running, the stance phase (Figure 1B) may also be divided into two subphases, absorption and propulsion (or generation), which are separated by midstance. In addition, Novacheck (1995, 1999) describes the middle of stance as stance phase reversal as the phase changes from force absorption to force generation at toe off.

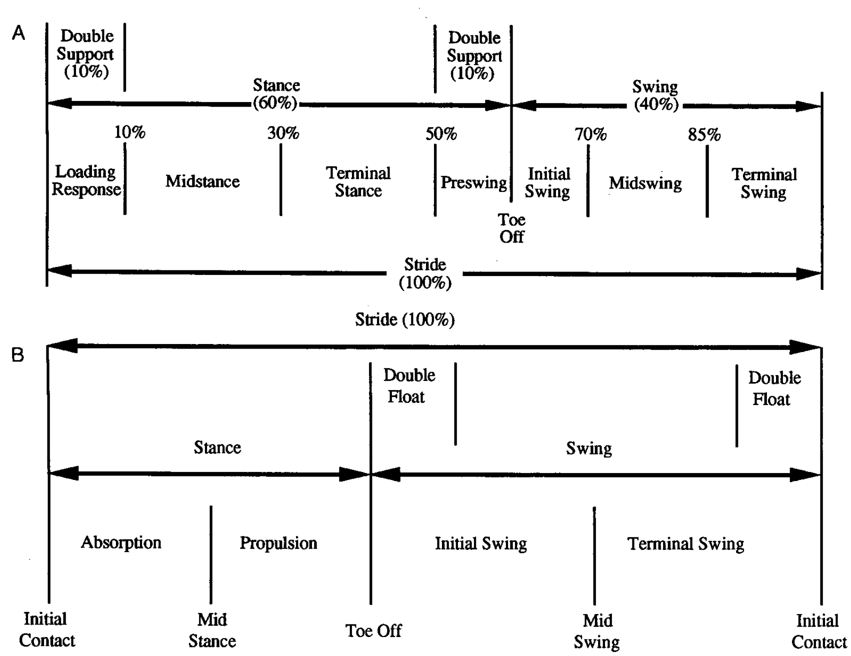


Figure 1. A comparison of the gait cycle terminology and phases used for (A) walking and (B) running.

From "The biomechanics of walking and running," by S. Ounpuu, 1994, *Clinics in Sports Medicine*, 13, p.845.

Normal walking can also be defined in terms of the double leg and single leg support times. Double support, when the two limbs are in contact with the ground, occurs at the first and last 10% of stance phase or loading response and preswing, respectively (Novacheck, 1995; Ounpuu, 1994). Single leg support, when one foot is in contact with the ground, is equal to the swing time of the opposite limb. Running differs from walking in that the two periods of double support are replaced by periods referred to as double float, when neither foot is on the ground. During running, stance time or foot contact time will always be less than the time of swing to accommodate these periods of double float (Gage, 1990). Therefore, the stance phase must represent less than 50% of the gait cycle and correspondingly, the swing phase must represent greater than 50% of the gait cycle (Ounpuu, 1994). Novacheck (1995, 1998) noted that as velocity increased, the relative length of stance decreased from 62% to 39% for running, while swing phase increased from 38% to 61%. Mann and Hagy (1980) also noted a stance phase of 62% for walking and 31% for running. Ounpuu (1990) reported a 43% stance phase with running, however, her participant group ran at a slower velocity than both Novacheck's and Mann's groups.

Further characterization of the gait cycle can be made with other temporal and stride variables such as velocity, cadence, step length, and stride length (Cavanagh, 1987; Gage, 1990; Ounpuu, 1994). Cadence is defined as the number of steps per unit of time (Ounpuu, 1994). Step length is the longitudinal distance between the two feet. Stride length extends from the initial contact of one foot to the following initial contact of the same foot, or, represents the sum of the right and left step lengths (Cavanagh, 1987; Gage, 1990). It should be noted that most of the temporal and stride variables are interrelated; as the speed of gait increases, step length, cadence, and velocity all increase (Mann & Hagy, 1980).

Kinematics

Kinematics are a description of movement and do not consider the forces that cause that movement. Joint kinematics refer to the variables that describe the spatial movement between segments such as joint angular motion measured in degrees (Ounpuu, 1994). Joint kinematic variables are frequently graphed as a function of the percentage of the gait cycle. While angle-angle diagrams are another common representation of joint kinematic data, these have less meaning for the practicing clinician and will not be considered here.

Most of the motion that occurs during walking occurs in the sagittal plane. In running, however, one generally observes greater range of motion (ROM) in all three planes of movement (sagittal, frontal/coronal, and transverse). Greater joint excursion has been noted in hip flexion, knee flexion, and ankle dorsiflexion with running (Dugan & Bhat, 2005; Mann & Hagy, 1980; Ounpuu, 1994). Other joints also exhibit a greater ROM, such as the pelvis and lumbar spine (Dugan & Bhat, 2005). In general, increasing the speed of running increases the amount of joint excursion, particularly in the sagittal plane (Dugan & Bhat, 2005; Mann & Hagy, 1980; Ounpuu, 1994).The mean joint kinematic patterns of movement for the pelvis, hip, knee and ankle during walking and running are presented in Figure 2. Joint rotations are shown for all three movement planes. A description of the findings relevant to the pelvis, hips and knees follows.

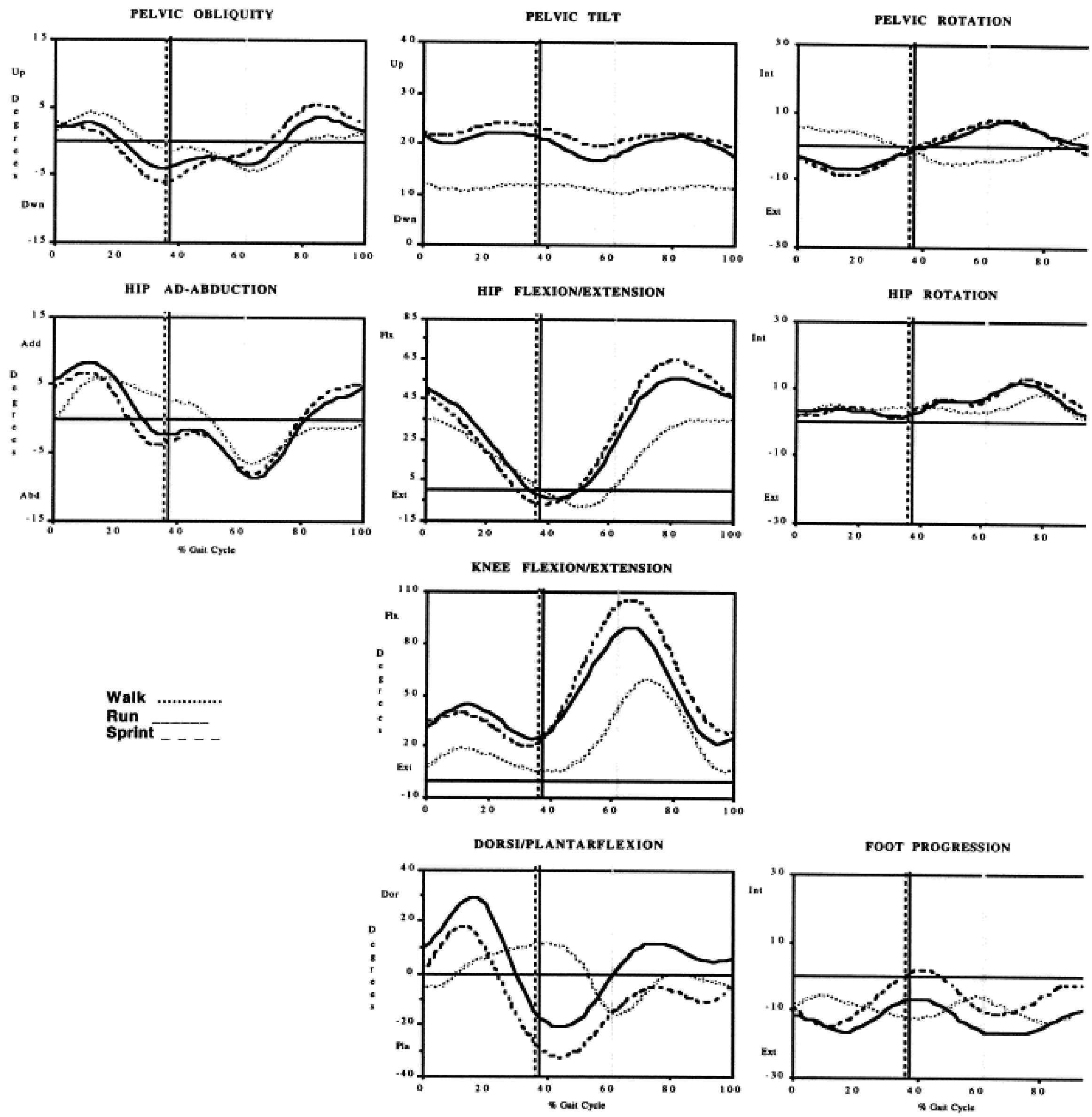


Figure 2. Kinematic results for walking, running, and sprinting. These graphs show the changing position of the joint listed for one complete gait cycle in all three planes. Each graph begins and ends at initial contact and therefore represents one gait cycle along the *x*-axis. The vertical dashed line represents toe off for each condition. The position of the joint or body segment in degrees is represented along the *y*-axis. The position of the pelvis is plotted relative to the horizontal and vertical coordinate system of the lab. Hip position represents the position of the femur plotted relative to the position of the pelvis.

From “The biomechanics of running” by T. Novacheck, 1998, *Gait & Posture, 7,* p. 81.

Sagittal plane kinematics. When observing sagittal plane motion there is a general shift into greater degrees of flexion at all joints during running. In addition, the center of mass is lowered as the motion changes from walking to running (Novacheck, 1998a). Rotation of the pelvis around a medial-lateral axis in the sagittal plane is known as anterior and posterior tilting (A-P) (Schache et al., 1999). The amplitude of this movement has been reported to range from 4-7 degrees (Ounpuu, 1994; Schache et al., 1999). The pattern of movement in the tilt of the pelvis is similar at all speeds (Figure 2). A-P tilting displays a biphasic curve during one gait cycle of walking or running (Ounpuu, 1990; Schache et al., 1999). The mean A-P tilt angle during running is usually between 15 and 20 degrees of anterior tilt. This is slightly greater than the degree of anterior tilt during normal standing which is found to be approximately 11 degrees (Schache et al., 1999). The pelvis is more anteriorly tilted as the speed of running increases. One might expect a greater amount of pelvic motion with faster velocities; however, A-P tilting appears to increase very little with faster running. It is believed that pelvic motion is minimized to conserve energy and maintain efficiency of movement (Novacheck, 1995; Novacheck, 1998a; Schache et al., 1999).

Sagittal plane hip motion follows a nearly sinusoidal pattern especially in walking. Maximum hip extension occurs just before toe-off and maximum flexion occurs in mid to terminal swing (Novacheck, 1995; Novacheck, 1998a; Schache et al., 1999). Maximum hip extension is similar in timing and magnitude during both walking and running but occurs just slightly later in the gait cycle (at the time of toe-off) during running (Novacheck, 1995; Novacheck, 1998a). As running speed increases, swing phase hip flexion increases leading to a longer step length; greater hip flexion would account for the increase in distance traveled with each step (Novacheck, 1995; Novacheck, 1998a). Unlike walking, the hip extends during the final third of swing phase in preparation for initial contact during running. It has been demonstrated that this hip extension during the final stages of terminal swing reduces the horizontal velocity of the foot prior to foot strike which would help to minimize the retarding ground reaction force at heel or foot strike (Novacheck, 1998a; Schache et al., 1999).

Although the pattern of sagittal plane knee motion in walking and running is very similar, the extremes of motion are very different. In running, during the absorption period of the stance phase, the knee flexes to approximately 40-45 degrees. This is followed by knee extension to an average of 20 - 25 degrees during the propulsion phase (Novacheck, 1995; Novacheck, 1998a; Ounpuu, 1990; Ounpuu, 1994). This progressive knee extension in the propulsion or generation period of the run cycle leads to greater knee extension which contributes to the increased step length on the contralateral side. This indicates that the knee is much more important in shock absorption during running (Novacheck, 1995). Sagittal plane knee joint ROM differences also exist between walking and running during swing phase. Maximum knee flexion during swing phase in normal walking is approximately 60 degrees (Novacheck, 1995; Novacheck, 1998a; Ounpuu, 1994). This is significantly different than the average of 90 degrees during running (Novacheck, 1998a).

Coronal plane kinematics. Overall, coronal or frontal plane motion is much more subtle than sagittal plane motion. Given that the only lower extremity joint with significant motion in the coronal plane is the hip, it is the one of most importance (Novacheck, 1995). Numerous studies have investigated the rotation of the hip about the anterior-posterior (A-P) axis (abduction-adduction) during running and have measured consistent results. All studies have measured the motion of the thigh segment in relation to the pelvis to obtain a hip angle. Hip angle amplitudes of 13 degrees have been reported during walking (Ounpuu, 1994). Hip angles during running are only slightly greater (14-16 degrees) (Novacheck, 1998a; Ounpuu, 1994; Schache et al., 1999). Generally, the hip is adducted in stance phase and abducted during swing phase. At foot strike, the hip is in an adducted position. During initial stance, when shock absorption is occurring, hip adduction increases slightly (~6 degrees) (Ounpuu, 1990). During the propulsive period of stance, the hip progressively abducts reaching a slightly abducted position by toe off. Further hip abduction continues during early swing. Maximal hip abduction of the swing leg occurs around midswing. During terminal swing the hip begins to adduct again. The hip abduction (~6 degrees) during terminal swing probably aids in clearance of the contralateral swinging limb (Ounpuu, 1990). The hip adduction during terminal swing possibly functions to position the lower limb for initial contact (Schache et al., 1999).

Rotation of the pelvis about an A-P axis during walking and running is known as pelvic obliquity or lateral pelvic tilt ((Schache et al., 1999). Variation exists between authors regarding the reported amplitude of pelvic obliquity during walking and running. Ounpuu (1994) reports 2 degrees of frontal plane motion during running and 8 degrees during walking. Novacheck (1998) reported 7-12 degrees based on velocity of running. Hip motion in the coronal plane mirrors pelvis motion to a great extent in such a way as to minimize the relative motion between the lower extremities and the body segments above (Novacheck, 1995). At foot strike the pelvis is obliquely aligned, being slightly higher on the stance (ipsilateral) side and slightly lower on the swing (contralateral) side (Novacheck, 1998a; Schache et al., 1999). As the limb is loaded, the pelvis continues to drop and the hip progressively adducts relative to the pelvis. Pelvic obliquity, with its associated hip adduction, is thought to play a role in shock absorption and in controlling the smooth descent and ascent of the body's center of gravity (Schache et al., 1999).This shock absorbing mechanism is similar to the role knee flexion plays in the sagittal plane (Novacheck, 1998a).

Coronal plane motion at the knee and ankle is anatomically limited by the collateral ligaments making them, in some authors' minds (Novacheck, 1995; Novacheck, 1998b; Ounpuu, 1990; Ounpuu, 1994), insignificant contributors of motion in this plane. However, numerous orthopedic physicians, physical therapists and athletic trainers believe that frontal plane motion at the hip/knee complex is significant, particularly from a functional perspective. Numerous studies (Dugan, 2005; Feller et al., 2007; Fulkerson, 2002; Messier et al., 1991) discuss the effect of the Q-angle as it relates to some orthopedic issues, particularly PFP. The Q-angle is a frontal plane kinematic measure. The Q-angle is defined as the angle formed by the intersection of a line from the anterior superior iliac spine (ASIS) of the pelvis to the midpoint of the patella and a line from the midpoint of the patella to the midpoint of the tibial tubercle; or the angle that results from the intersection of the axis of pull of the quadriceps muscle and the axis of the patellar tendon (Feller et al., 2007). Physicians and clinicians have noted a "medial collapse" of the knee during functional activities such as jumping and running (Feller et al., 2007). This collapse can be measured by "the so-called dynamic (functional) Q-angle" (p. 547) (Feller et al., 2007). The authors note a conundrum of whether the dynamic Q-angle is determined from the hip down or from the foot up (Feller et al., 2007). “The former suggests that increased femoral adduction and internal rotation at the hip are fundamental to an increased dynamic Q-angle. The foot-up model is based on the premise that increased foot pronation results in tibial internal rotation producing an in-turned knee posture and “functional” valgus” (p. 547). It has also been postulated that increased anterior pelvic tilt may have an effect of increasing dynamic valgus at the knee and therefore the dynamic Q-angle, by causing relative internal rotation of the femur (Arendt, 2000; Feller et al., 2007; Hruska, 1998).

Transverse plane kinematics. Motion in the transverse plane, as in the coronal plane, is small in magnitude compared to the sagittal plane. Joint rotations may be difficult to comprehend because they are difficult to see and harder to measure; measurement variability is, many times, greater than the mean measure itself (Reischl, Powers, Rao, & Perry, 1999). The movement patterns in the transverse plane, however, are important for energy efficiency and are equally important for single leg stability during stance and during gait. It is this author’s opinion that transverse plane movements can be the most difficult to retrain during rehabilitation efforts and are the movements most neglected or under-rehabilitated in physical therapy/athletic training programs.

Rotation of the pelvis about a vertical axis in the transverse plane is known as axial rotation or internal and external rotation. Internal pelvic rotation occurs when the reference side of the pelvis is anterior and external pelvic rotation occurs when the reference side of the pelvis is posterior (Schache et al., 1999). Stated another way, these rotations show the relationship of the pelvis with respect to the direction of progression.

For example, an internally rotated left hemipelvis indicates that the left side is toward the direction of progression (Ounpuu, 1990). Foot rotations are presented in the same way. Hip rotations, however, are shown in relation to the position of the pelvis. The amplitude of pelvic movement in this plane has been reported to be 8 degrees during walking (Ounpuu, 1994) and between 16 and 18 degrees during running (Novacheck, 1998a; Ounpuu, 1994; Schache et al., 1999).

The function and motion of the pelvis in the transverse plane is very different during walking and running; pelvic motion during running is interestingly opposite that of walking. In walking, the pelvis is internally rotated in stance phase to increase step length. The pelvis is not utilized as a stride lengthener during running. During running, the pelvis externally rotates on the side of the lower limb preparing for foot strike. By foot strike the pelvis is slightly externally rotated on the stance side, which continues to increase until a maximal position of external rotation is reached around midstance (Schache et al., 1999). During terminal stance, the pelvis begins to internally rotate on the stance side such that by toe-off the pelvis is in a neutral position. Internal rotation on the swing side continues through the early swing period, reaching maximum internal rotation around midswing. The pelvis then begins to externally rotate again on the side of the lower limb in terminal swing (Novacheck, 1998a; Schache et al., 1999). It is believed that this pattern of pelvic rotation during running is important for energy efficiency (Novacheck, 1998a; Schache et al., 1999). For example, an externally rotated pelvis on the stance leg may actually assist in decreasing the posterior component of the ground reaction force (Schache et al., 1999). This maximizes the horizontal propulsion force and helps avoid potential loss of speed during running (Novacheck, 1998a).

Several authors have provided kinematic descriptions of transverse plane hip joint rotation during running (James & Brubaker, 1973; Mann, 1982; Mann, 1989; Slocum & James, 1968). At foot strike the hip joint is thought to be externally rotated. During the absorption period of stance, the hip joint then internally rotates (James & Brubaker, 1973; Mann, 1982; Mann, 1989; Slocum & James, 1968). There is disagreement between the authors regarding the propulsion period of stance. James and Brubaker (1973) and Slocum and James (1968) described the hip to continue internally rotating, while Mann (1982, 1989) believed that progressive external rotation occurred. Vertical axis rotation of the hip joint has been relatively well documented in walking ((Mann, 1982; Mann, 1989). However, when a person starts to run the quantitative measurement of this motion becomes difficult to obtain (Schache et al., 1999). Ounpuu (1994) and Novacheck (1998) measured transverse plane hip motion during running. Both found the hip joint to internally rotate during the absorption period of stance, after which the hip returned to a neutrally rotated position by toe-off (Novacheck, 1998a; Ounpuu, 1994). Ounpuu (1994) found the hip to remain in a neutrally rotated position, while Novacheck (1998) found an increase in internal rotation during mid swing before returning to a more neutral position by terminal swing. When interpreting kinematic data in the transverse plane, one needs to keep in mind the significant degree of error that may occur with measurements obtained utilizing skin markers ((Reinschmidt, van Den Bogert, Murphy, Lundberg, & Nigg, 1997; Reinschmidt, van den Bogert, Nigg, Lundberg, & Murphy, 1997).

Kinetics

Kinetics involves the study of forces that produce movement, such as ground reaction forces, joint moments and powers. Other kinetic measures of interest during human locomotion include the assessment of center of pressure (COP) and the mapping of pressure distributions at the foot. Compared with kinematics, which are purely descriptive, kinetics may provide more concrete information into the cause of movement and movement abnormalities (Ounpuu, 1990).

Force measurements. Force platforms are commonly found in gait laboratories as they are the "gold standard" for describing the patterns of ground reaction forces during walking and running. However, they do have their limitations when attempting to assess function not only during walking and running, but more specifically during attempts to assess more functional activities, such as clinical tests and specific sports activities. In-shoe pressure distribution systems, such as the Emed sensor platform and the Pedar insole system (©novel electronics inc., St. Paul, MN) are frequently used in clinical and research environments. The Pedar system, commonly used in sport-related gait assessments and clinical weight-bearing activities, is a vertical pressure measuring device. Pedar has been used frequently to measure vertical force components during activities when a force platform was unavailable or inconvenient. The validity and reliability of the Pedar system have been previously documented (Barnett, Cunningham, & West, 2001; Hayes & Seitz, 1997; Hurkmans, Bussmann, Benda, Verhaar, & Stam, 2006; Kalpen & Seitz, 1994; Kernozek, LaMott, & Dancisak, 1996; Konczak & Anderson, 2000; Orlin & McPoil, 2000; VanZant, McPoil, & Cornwall, 2001). Test-retest reliability ICC's ranged from 0.84-0.99 depending on variable analyzed. The Gait and Posture Laboratory at this Institution has confirmed the manufacturer's claims that the Pedar system does measure vertical force output to within ±5% of the actual force measured by a force platform (Konczak & Anderson, 2000).

The ground reaction force (GRF) has three component forces: vertical, anterior-posterior or fore-aft, and medial-lateral. The GRF point of action is at the center of pressure on the bottom of the foot (Ounpuu, 1990). The vertical force component provides information on the way in which the body is cushioned during and after foot strike and then again when the body is accelerated upward into the flight phase of running or during toe-off in walking (Cavanagh, 1987). This component, the largest component, resembles the resultant GRF in shape and magnitude (Ounpuu, 1990). During walking, vertical ground reaction forces reach magnitudes that are 1.1 to 1.5 times body weight (depending on walking speed) during loading response and again at toe-off (Ounpuu, 1990; Ounpuu, 1994; Rogers, 1988). The peak vertical force registered in stance during jogging or long distance running is 2 to 3 times that of walking (Mann, Baxter, & Lutter, 1981; Ounpuu, 1990; Ounpuu, 1994; Perry, 1990).

The anterior-posterior or fore-aft ground reaction forces during walking represent braking forces during the first 50% of stance followed by a propulsive generation during the last 50% (Ounpuu, 1990). The peak amplitudes of these forces are about 30% of body weight. During running, an additional peak braking force occurs at approximately 10% of stance. The braking-propulsion ratio is approximately the same in walking and running, with average magnitudes of 20% of body weight (Mann et al., 1981; Ounpuu, 1990). While Mann (1981) and Ounpuu (1990) agree that the braking-propulsion ratio is similar in walking and running, they disagree regarding the magnitude of the A-P force component. Ounpuu reports similar fore-aft forces during both activities, while Mann describes forces that are 50% greater in magnitude during running. However, although not assessed in her particular study, Ounpuu (1990) stated that there is evidence that running velocity does affect the magnitude of the GRF; it is generally concluded that all vertical GRF variables increase with running speed.

Medial-lateral shear forces represent how the center of mass of the body is transferred from side to side during ground contact. They are of minimal magnitude during walking, jogging and running and represent only 10-25% of body weight (Ounpuu, 1990). It is not uncommon to see differences between the left and right foot of the same individual in all ground reaction force measures (Cavanagh, 1987), however, some believe the variability may be more apparent in the medial-lateral component. Some authors have utilized this component as a measure of stability during gait (Mann et al., 1981; McClay & Manal, 1999).

The rates at which forces and pressures are applied to the body have also been studied. Vertical loading rates for running have been explored (Munro, Miller, & Fuglevand, 1987); however, correlations to specific injuries have not been substantiated. Although GRF are thought to play a role in running injuries, it is still unknown whether it is peak forces, loading rate, number of repetitions, or some combination of factors that places a runner at risk for injury (McClay, 2000).

Center of Pressure (COP) measurements. Another method of evaluating force application to the foot and, ultimately, the lower extremity, is the assessment of center of pressure and pressure distribution data. The COP pathway obtained from a force plate or pressure distribution system during the ground contact phase of locomotion provides information on the path of the instantaneous point of application of the resultant foot-floor vertical ground reaction vector (Katoh, Chao, Laughman, Schneider, & Morrey, 1983). At a specific point in time, the COP indicates the unique point defined such that the net rotational torque resulting from all forces and pressures applied medially and laterally to the point is zero (Chesnin, Selby-Silverstein, & Besser, 2000). The path of this moving foot pressure center produces a characteristic pattern, which Katoh et al. (1983) call the COP pattern. The COP pattern has frequently been used as a measure of balance to quantify postural control during standing (Adkin, Frank, Carpenter, & Peysar, 2000; Carpenter, Frank, Winter, & Peysar, 2001; Winter, 1995; Wu & Chiang, 1996) and balance control during locomotion (Gefen, Megido-Ravid, Itzchak, & Arcan, 2002; Han, Paik, & Im, 1999). COP measures quantified using a force platform only provide resultant force vector information and point of application. There is a lack of information provided on the location of the foot relative to the applied forces. This is a limiting factor when studying standing balance and gait stability (Dixon, 2006). Pressure distribution systems quantify the distribution of forces across the plantar surface of the foot or shoe, providing a simultaneous record of the footprint and the COP. Pedar terms this recorded COP pattern the gait line. COP/gait lines are graphically represented in many ways. Pedar software provides one representative method as seen in Figure 3. The right foot demonstrates a normal representation of the COP path which originates just slightly lateral to the midline of the heel, travels along the midline of the foot, and up to the metatarsal heads.

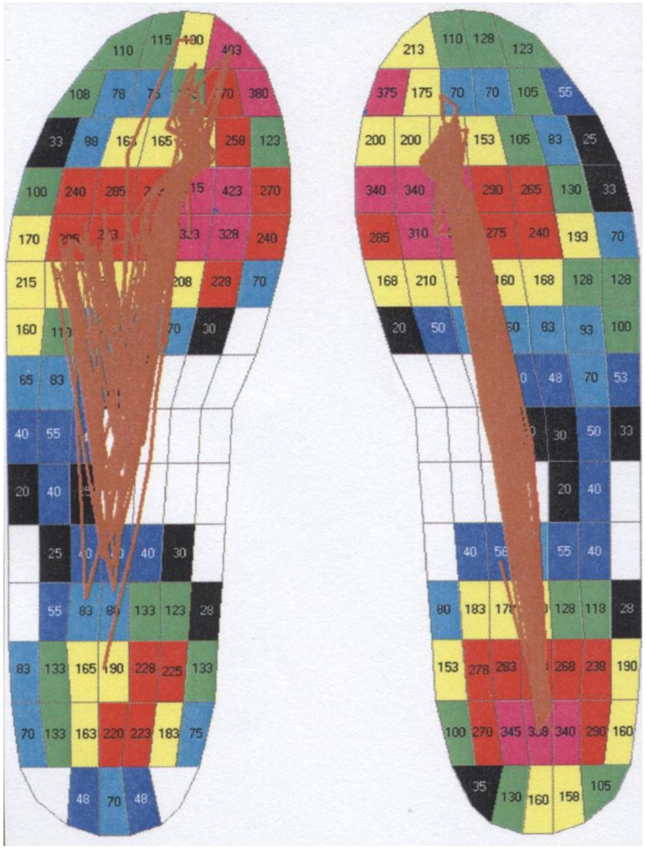


Figure 3. Graphic representations of gait lines from one subject following a 30 second Pedar run trial.

Pressure migrates medially so that by toe-off the COP lies under the first or second toe. This medial migration aspect of the COP path has been described as the most variable among subjects (Rogers, 1988).

During studies of walking, the mediolateral stability under different conditions has been represented using a mediolateral deviation of the COP. This has allowed for the study of footwear and shoe insert effects on balance during locomotion. Gefen et al. (2002) and Han et al. (1991) used mediolateral deviations in the COP to quantify stability when walking in high-heeled shoes. Katoh and colleagues (1983) looked at how the COP path was altered by different footwear. Although the COP has been assumed to provide information on the mediolateral balance of the foot, the relationship between observed differences in the COP and inversion/eversion movement of the foot has not always been clear. Numerous authors have looked at the effects of foot inserts, orthotics and posting mechanisms on inversion/eversion movements of the foot (Dixon, 2006; Nigg et al., 2003; Van Gheluwe & Dananberg, 2004). Studies of the COP and balance have generally focused on walking; studies of running COP as a measure of balance are less common. Despite this lack of scientific evidence, pressure systems are increasingly used as a tool for assessment of footwear in runners.

Joint Moments and Power

Two additional common measures used to describe locomotion kinetically are joint moments and joint power. For a detailed description of these results the reader is referred to Ounpuu's chapter in *Instructional Course Lectures (Volume 39)* (Ounpuu, 1990) and Novacheck's (1998) review paper on the biomechanics of running. Their descriptions rely heavily on sagittal plane information. Until recently, little has been known about lower extremity joint kinetics in the secondary planes of movement. However, many of the proposed mechanisms of injury occur in the frontal and transverse planes of movement. McClay and Manal (1999) have reported on 3-D kinetics of the rearfoot and knee during running. The results do suggest that a significant amount of work is done in the frontal and transverse planes of motion. As with kinematic measures, transverse plane motion does exhibit the greatest amount of between-subject variability. Therefore, kinetics in this plane of motion should be interpreted with caution.

Gender Differences in Lower Extremity Biomechanics

Walking

There is a common perception that males and females walk differently. For example, observers can consistently identify the gender of a person based solely on the dynamic light displays of the subject's joints (as measured by markers) while walking in the sagittal plane (Cutting, Proffitt, & Kozlowski, 1978). Despite these recognized perceptions of gender differences, there are only a few studies documenting specific biomechanical differences about the pelvis, hip and knee during walking (Kerrigan et al., 1998; Schache et al., 2003; Smith et al., 2002). The literature does provide information regarding temporal gait parameter differences between males and females. On average, women walk at significantly higher cadences and with shorter step lengths than men (Kerrigan et al., 1998; Oberg, Karsznia, & Oberg, 1993; Smith et al., 2002). However, when normalized for height, women tend to have the same or slightly greater stride lengths as compared to men (Kerrigan et al., 1998; Smith et al., 2002). Despite these temporal parameter differences, females and males generally tend to walk at the same velocity (Kerrigan et al., 1998; Oberg et al., 1993; Smith et al., 2002).

Gender also appears to be a factor influencing kinematic and kinetic patterns. Chiu and Wang (2007) recently reported significantly higher muscle activity in the tibialis anterior with increased ankle joint motion in women. They also found a greater vertical ground reaction force during loading response and preswing phases in women as compared to men. Others have replicated the gender difference in vertical GRF during walking (Ferber et al., 2003). Kerrigan et al. (1998) found that females had significantly greater hip flexion and less knee extension before initial contact, greater knee flexion moments in preswing and greater peak mechanical joint power absorption at the knee during preswing. Smith et al. (2002) looked at pelvic motion gender stereotypes and found that women had a significantly greater range of pelvic obliquity and significantly lower normalized vertical center of mass (COM) displacement. Pelvic obliquity motions are linearly related to lower lumbar spine movement, implying that greater lumbosacral motion may be implicated in the progression of lumbar disc disease (Smith et al., 2002). However, a reduction in vertical COM displacement implies a more biomechanically efficient gait, at least in terms of the work required to lift the COM during walking (Smith et al., 2002). Thus, greater pelvic obliquity in women may represent a mechanism to reduce vertical COM displacement and conserve energy during walking (Smith et al., 2002).

Running

Women runners are reported to be twice as likely to sustain certain running injuries such as PFP, iliotibial band friction syndrome, and tibial stress factures as compared to men (Taunton et al., 2002). It has been suggested that known differences in structure (skeletal alignment, muscle strength and anthropometric parameters) may predispose females to differences in running mechanics, which may lead to specific injuries. While structural pelvifemoral gender differences have been studied, little attention has been given to running patterns between men and women.

Structurally, Horton and Hall (1989) refuted the conventional notion that women have a wider pelvis than men. However, they do report that women have a larger hip width to femoral length ratio which leads to greater hip adduction in standing and during gait. This associated increased angulation of the femur contributes to the greater genu valgus knee position frequently reported in women (Ferber et al., 2003). Women have also been shown to exhibit greater active hip internal rotation than men (Simoneau, Hoenig, Lepley, & Papanek, 1998). The structural combination of increased hip adduction, hip internal rotation, and genu valgus may explain, in part, the larger Q-angle in women (Feller et al., 2007; Ferber et al., 2003; Horton & Hall, 1989). An increased Q-angle has been shown to be associated with an increase in lateral patellar contact forces (Feller et al., 2007; Fulkerson, 2002) and hence, may play a partial role in the greater incidence of PFP in women (Feller et al., 2007; Ferber et al., 2003; Messier et al., 1991).

Three studies have been found that address differences in pelvifemoral/lower extremity joint mechanics between genders during running. Malinzak, Colby, Kirkendall, Yu, & Garrett (2001) studied the sagittal and frontal plane motion of the knee in male and female runners. They reported that while frontal plane excursion was similar in males and females, females exhibited 11° more knee valgus motion throughout the stance phase. In addition, females exhibited less peak knee flexion and less knee flexion excursion compared to men. Unfortunately, these authors did not examine hip kinematics in these subjects.

Ferber et al. (2003) and Schache et al. (2003) looked at all 3 planes of motion in their running studies. In addition, Schache and colleagues investigated anthropometric and spatio-temporal parameters between genders. Women were significantly shorter and displayed significantly greater standing pelvic tilt angles. Women also displayed a shorter stance time, swing time, stride time and stride length; women also had a higher stride cadence. Ferber et al. (2003) found no difference in stance time or relative stride lengths between genders.

Females were found to run with a significantly increased mean position of anterior pelvic tilt during the running cycle compared with males (Schache et al., 2003). This result may be the product of the greater standing anterior pelvic tilt angle in female runners in this study. Females also tend to have a greater angle of peak hip flexion during late swing phase of running and displayed significantly greater amplitudes of pelvic motion (all 3 planes) and hip adduction-abduction motion. Schache, Blanch and Murphy (2000) found that anterior pelvic tilt and hip extension are coordinated movements during running. They reported that anterior pelvic tilt tended to be increased in runners who displayed reduced absolute peak hip extension ROM during terminal stance. Ferber et al. (2003) refuted Malinzak's (2001) results, finding no significant differences in sagittal plane hip and knee kinematics and kinetics. However, females were found to exhibit significantly greater peak hip adduction angle, greater hip adduction velocity and greater hip frontal plane negative work compared to men. It is concluded that greater eccentric demands may be placed on the hip abductors of women compared to men (Ferber et al., 2003). In the transverse plane, female runners exhibited greater hip internal rotation at heel strike resulting in greater external rotation excursion and velocity compared to male runners. In conclusion, female runners appear to exhibit significantly different hip and knee kinematic and kinetic patterns compared to men, particularly in the secondary planes of motion.

Influence of Altered Lower Extremity Kinematics and Kinetics on Patellofemoral Dysfunction

The clinical diagnosis of PFP typically encompasses anterior or retropatellar knee pain that is aggravated by some combination of activities that load the patellofemoral joint (i.e. prolonged sitting, stair ascent or descent, squatting, kneeling, running and jumping). A commonly accepted hypothesis of the cause of PFP is that abnormal patellar tracking increases patellofemoral joint stress and causes subsequent wear on articular cartilage surfaces and/or dysfunction within the patellofemoral joint space (Feller et al., 2007; Robinson & Nee, 2007). Patellar malalignment and/or abnormal patellar tracking is thought to be one of the primary precursors of patellofemoral joint pathology (Powers, 2003). Thus, numerous studies have taken this more localized approach to PFP, investigating patella kinematics and patellofemoral contact pressures alone and in conjunction with lower extremity joint kinematics (Amis, Senavongse, & Bull, 2006; Bennell, Duncan, & Cowan, 2006; Brechter & Powers, 2002a; Brechter & Powers, 2002b; Hreljac et al., 2000; Lee, Morris, & Csintalan, 2003; Li, DeFrate, Zayontz, Park, & Gill, 2004; MacIntyre et al., 2006). Interventions based on this approach have had mixed results. One suggested reason for these relatively poor outcomes may be related to the assumption that there is instability of the patella and that instability is the result of the patella moving on the femur (Mascal et al., 2003). Recent evidence suggests that PFP pathology may be the result of the femur rotating underneath the patella in the transverse plane and that the patellofemoral joint is dramatically influenced by the segmental influence of the pelvis and lower extremity (Feller et al., 2007; Hruska, 1998; Mascal et al., 2003; Powers, 2003).

Numerous authors now believe that the patellofemoral joint needs to be assessed in the context of the whole lower limb and the whole body (Feller et al., 2007; Hruska, 1998). Hruska (1998) suggests that pelvic position is the foremost contributing influence on patellofemoral joint biomechanics. He suggests that many patients with PFP present with an anteriorly tilted pelvis. Femoral rotational direction is guided and limited by the rotation and tilt of the pelvis and by soft tissue structures at the hip. “An anterior tilt or rotation of the pelvis promotes overuse of the hip flexors (i.e. there is adaptive shortening of the ilopsoas and iliotibial band and increased tonic activity within the tensor fascia latae and iliopsoas) during beginning stance, and swing phases of gait, as well as forward pelvic rotation, femoral internal rotation, medial displacement of the femoral range of rotation, genu valgus, genu recurvatum, subtalar eversion, and forefoot or rearfoot pronation (Figure 4). This sequence of displacements leads to pathokinesiologic effects on the knee” (p. 23) (Hruska, 1998). The increased femoral internal rotation would result in excessive pressure on the lateral aspect of the patella; other muscular strength and flexibility issues result secondary to these positional changes.

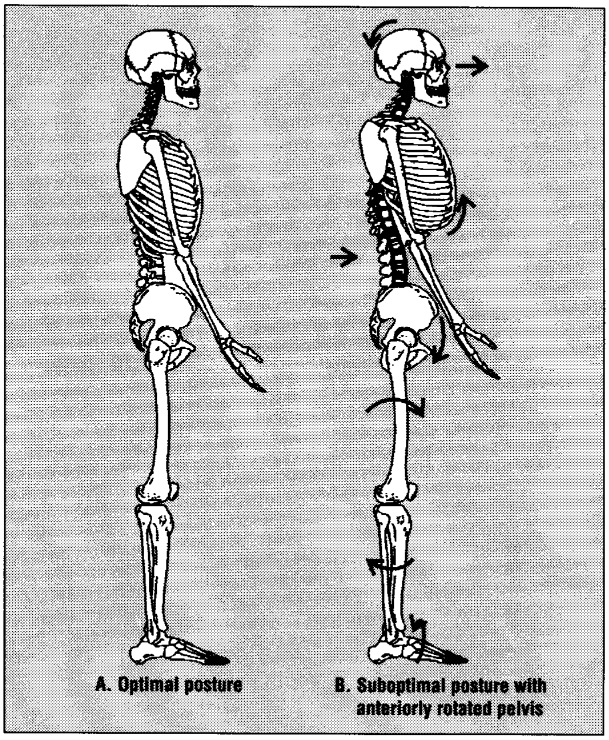


Figure 4. Schematic illustration of the sequence of pelvifemoral kinematic displacements associated with an anterior pelvic tilt.

From "Pelvic stability influences lower-extremity kinematics," by Hruska, R., 1998, *Biomechanics*, 10, p. 24. Reprinted with permission of the author.

There is increasing evidence that patients with PFP have hip muscle weakness (Cichanowski et al., 2007; Ireland et al., 2003; Mascal et al., 2003; Niemuth et al., 2005; Robinson & Nee, 2007). Hip external rotation and abduction musculature contribute to pelvic stability and lower extremity alignment by controlling femoral internal rotation and influencing hip adduction during weight-bearing activities (Hruska, 1998; Hruska & Joutras, 1999; Piva et al., 2005). It is suggested that weakness of these muscles may increase femoral internal rotation, valgus knee moments, or cause lateral hip dysfunction during gait (Hruska & Joutras, 1999; Piva et al., 2005).These musculoskeletal dysfunctions may alter the hip abduction/adduction moments at the hip or lead to an increased Q-angle, which may subsequently alter patellar tracking, leading to knee pain (Hruska & Joutras, 1999; Piva et al., 2005).

In addition, these same authors have shown additional significant hip strength deficits in subjects with PFP. When compared against controls, subjects with PFP show significant deficits in hip abduction, hip extension, and hip external rotation strength (Ireland et al., 2003; Mascal et al., 2003; Piva et al., 2005; Robinson & Nee, 2007; Willson & Davis, 2006). Weakness patterns in more proximal joints may affect more distal joint function. Stefanyshyn and colleagues (2006) looked at knee angular impulse as a predictor of PFP. They demonstrated that PFP patients had significantly higher knee abduction impulses than asymptomatic patients. Abduction moments represent the torque or rotational loads on the knee in the frontal plane. This increased impulse measure indicates associated dysfunction of secondary plane motion in PFP subjects.

Few studies exist that describe 3-D kinematic and kinetic results comparing subjects with PFP and controls during walking and running. Willson and Davis (2006), with a sample of 6 subjects, reported that females with PFP tend to demonstrate greater peak hip internal rotation and hip adduction during running than healthy control subjects. Mascal and associates (2003), in 2 case reports, also demonstrated excessive hip adduction, internal rotation, and knee valgus during walking and during a step-down maneuver. In addition, they found normal sagittal plane motion at the hip, knee and ankle. Results from both studies appear to support Hruska’s (1998) pelvifemoral model; however, both should be viewed with caution secondary to the small sample sizes.

Powers, Chen, Reischl and Perry (2002) compared the effects of foot pronation on tibial and femoral rotation in subjects with and without PFP. They had hypothesized, based on a theoretical paper by Tiberio (1987) that subjects with PFP would exhibit larger degrees of foot pronation, tibial internal rotation, and femoral internal rotation compared to individuals without PFP. They found no significant difference in the magnitude and timing of peak foot pronation and tibial internal rotation between the groups (n=30). Their only significant finding with respect to lower extremity kinematics was found at the femur. On average, the PFP group exhibited 2.1 degrees of external rotation, which was significantly different than the 1.6 degrees of internal rotation found in the control group (Powers et al., 2002). Peak femur rotation was found to occur significantly later in the PFP group. They hypothesized that the reduced internal rotation found in the PFP subjects may have been a compensatory strategy to avoid pain (Powers et al., 2002).

Gait studies regarding PFP and kinetics are equally sparse. It has been suggested that patients with PFP may employ compensatory strategies at the knee to minimize symptom aggravation and reduce patellofemoral joint reaction forces during activities and gait (Salsich, Brechter, & Powers, 2001). Nadeau et al. (1997) observed that subjects with PFP had decreased knee flexion and reduced peak knee extensor moments during the stance phase of gait; however, again, their sample size was small (N=5). Powers et al. (1999) reported that subjects with PFP adopted slower walking velocity and demonstrated diminished quadriceps muscle activity compared to controls during level and ramp walking. In another study, this group also found reduced peak knee extensor moments during stair ascent and descent (Salsich et al., 2001). These findings may be suggestive of a “quadriceps avoidance” gait pattern (Salsich et al., 2001).

In studies investigating ground reaction forces during gait, it has been reported that the average vertical, anterior-posterior and mediolateral forces produced by PFP subjects were lower than those of a control group (Callaghan & Baltzopoulos, 1994). The average peak loading rate for a PFP group was also found to be significantly less than the control group in self-selected and fast walking (Powers et al., 1999). While it is evident that kinematic and kinetic changes do occur at the knee in subjects with PFP, these early studies predominantly explore sagittal plane motion. Exploration of kinematic and kinetic dysfunction in the secondary planes of motion is more limited.

Symmetry and Limb Laterality

Gait symmetry has been defined as a perfect agreement between the actions of the lower limbs, suggesting that there are no statistical differences noted on parameters measured bilaterally (Goble, Marino, & Potvin, 2003; Herzog, Nigg, Read, & Olsson, 1989; Sadeghi, Allard, Prince, & Labelle, 2000). The presence of symmetry between the left and right lower limbs of a normal individual during walking is a common assumption made frequently in both clinical and research settings (Goble et al., 2003). Historically, gait symmetry was assumed in research for the sake of simplicity in data collection and analysis; many studies relied on unilateral limb data collection or pooled data between right and left lower extremities (Sadeghi et al., 2000). Another common justification for the assumption of symmetry in human gait has been that it is a necessary means of maximizing energy efficiency (Goble et al., 2003). However, this philosophy has been challenged based on studies of clinical populations with neurological asymmetry (i.e. hemiparetic stroke, unilateral amputation, etc.). In these populations, affected individuals preferably change their walking style to one of greater asymmetry in order to accommodate their physical limitations. This new pattern of gait, though markedly asymmetrical at times, usually has greater metabolic efficiency than one where greater symmetry is physically imposed on the subject (Goble et al., 2003). Mattes, Martin, and Royer (2000) clearly showed this in their study with amputees. They demonstrated that patterning a prosthetic limb to have more symmetrical weight and moment of inertia characteristics to that of the unaffected leg actually decreased energy efficiency of that individual's gait. This result suggested that optimizing the efficiency of gait is not necessarily the key determinant of symmetry in an individual, but rather, symmetry is the consequence of other factors (Goble et al., 2003).

Early evidence of gait symmetry utilizing bilateral limb data and 3-D electrogoniometry has been reported. Hannah, Morrison and Chapman (1984) demonstrated joint motion symmetry in all three planes of motion of the hip and in the sagittal plane of the knee during walking using time and frequency domain analysis. Hamill, Bates and Knutzen (1984) found no significant differences between limbs in 11 vertical, 5 anterior-posterior and 4 mediolateral characteristics of ground reaction forces during walking and running. These results were further supported by Menard et al. (1992) who found symmetry in all ground reaction force patterns observed during self-selected walking speeds in nine able-bodied subjects. Symmetrical electromyography (EMG) outcomes had been presumed for homologous muscles during walking, however, objective documentation was not provided early on. Although in a few studies EMG data were collected from the dominant limb, in many others data were collected only from the right or left lower limb. Arsenault et al. (1986) pooled the data across their subjects and reported nearly perfect symmetry of muscle profiles for muscle activities for individual subjects and found profiles to be highly repeatable within subjects. The investigators acknowledged, however, that pooling data smoothes out individual participation and that these overall averages cannot be perceived as the real profile of EMG activity (Arsenault, Winter, & Marteniuk, 1986)

Although it seems that gait symmetry has been assumed for simplicity of data collection and analysis, gait symmetry was actually only reported in a few studies using quantitative biomechanical data and using both limbs for evaluation (Sadeghi et al., 2000). Unfortunately, small sample sizes, unclear definitions of symmetry, and overly simplistic statistical methodology were limitations of these studies (Sadeghi et al., 2000).

The etiology of running injuries continues to puzzle both researchers and clinicians. In a given year, 50% of all runners will sustain a musculoskeletal injury and will subsequently be 50% more likely to become reinjured (Zifchock, Davis, & Hamill, 2006). The tendency of a runner to become injured on a certain side has been proposed to relate to lower extremity asymmetry. High levels of asymmetry are typically thought to be associated with pathology and marked differences have been noted between the affected and unaffected limbs of patients or subjects (Sadeghi et al., 2000; Zifchock et al., 2006). Examples such as leg length discrepancies, injury, and compensation for injury have all been associated with gait asymmetry (Perttunen, Anttila, Sodergard, Merikanto, & Komi, 2004; Williams, Cavanagh, & Ziff, 1987; Zifchock et al., 2006). It is believed that left untreated, significant asymmetry could lead to new or recurring overuse injuries. Therefore, clinicians often use restoration of symmetry as a goal for their treatment approaches (Robinson, Herzog, & Nigg, 1987; Zifchock et al., 2006). However, should full restoration of symmetry be the ultimate goal if frequent differences have been reported between the right and left limbs of able-bodied individuals? Instead, is there a threshold level of asymmetry above which injury may occur?

Singh (1970) and DuChatinier and Rozendal (1970) claimed that the lower limbs are not used equally during walking. Asymmetrical behavior of the lower extremities has been observed in spatio-temporal and kinematic parameters such as velocity profiles, step and stride length, foot placement angle, maximum knee flexion and range of joint motion (Gundersen et al., 1989; Sadeghi et al., 2000). However, although spatio-temporal parameters may provide an overall impression of gait, they may not provide insight into the foundation of the symmetrical or asymmetrical behavior. Thus, gait asymmetry has also been addressed examining kinetic data (Goble et al., 2003; Hamill et al., 1984; Herzog et al., 1989; Matsusaka et al., 1985; Sadeghi et al., 2000).

Asymmetry in gait is often described kinematically or kinetically as a difference between the right and left sides of the body. This is most often documented by calculating a symmetry index (SI) (Sadeghi et al., 2000; Zifchock et al., 2006). Robinson et al. (1987) originally proposed the SI as a quantitative indicator of the percent difference between a kinetic or kinematic parameter measured on the right and left sides. A value of zero indicates perfect symmetry, while increasing values (either positive or negative) indicate greater levels of asymmetry. Normal levels of asymmetry in gait mechanics appear to vary widely between parameters. Herzog et al. (1989) calculated SI values for kinetic variables in healthy subjects during walking. They found values that ranged from 0.1% for braking impulse to 711.1% for total anterior-posterior impulse. In a larger study, Williams et al. (1987) reported SI values for kinetic parameters measured in elite female runners. They reported side to side differences that ranged from 3.9% for peak vertical GRF to 28.3% for change in lateral velocity. Karamanidis et al. (2003) reported values ranging from 3.0% for knee joint angle at touchdown to 53.8% for hip angle velocity. Finally, Giakas and Baltzopoulos (1997) investigated the variability and symmetry of GRF measurements during walking utilizing time and frequency domain analysis. This study confirmed that human gait is a symmetrical movement based on harmonic analysis; however, substantial asymmetries characterized time domain variables particularly in the medio-lateral component of GRF (Sadeghi et al., 2000). All the studies involved runners who were currently healthy; no history of previous injuries was provided. In summary, based on these observations, it may be concluded that gait is asymmetrical (Sadeghi et al., 2000).

Sadeghi et al. (2000) argued whether it is acceptable to conclude that able-bodied gait is asymmetrical just because of the existence of statistically significant differences between corresponding parameters measured from the right and left limbs (termed local asymmetry)? If this is the case, how can one explain the effect of other factors such as compensation and/or adaptation that directly influence lower limb behavior during gait (Sadeghi et al., 2000)? Compensations occur in circumstances of pathologic gait. These may cause gait asymmetries when a parameter is compared with its corresponding value on the uninvolved side. This asymmetrical behavior is most likely related to the compensation rather than purely asymmetrical behavior of the lower limbs. Since compensations might also occur in able-bodied gait, recent gait studies have attempted to explain asymmetry as a functional behavior of lower extremities.

The aim of locomotion is to propel the body forward while supporting the body against gravity. It requires precise coordination between two tasks: propulsion and balance and/or control (Sadeghi, Allard, & Duhaime, 1997; Sadeghi et al., 2000). Hirasawa (1981) was among the first to interpret gait asymmetry in able-bodied subjects based on the support and mobility associated with each limb. He claimed that the left and right lower limbs have a supporting and moving function, respectively. While evaluating GRF components during walking, Matsusaka et al. (1985) reported that medio-lateral balance was mostly controlled by the left limb. In a large walking study, Hirokawa (1989) associated propulsion with the right limb while the left was found to be responsible for support. Later, Sadeghi et al. (1997) reported that propulsion was related to the leg with predominant muscle power generation whereas support and control functions were associated with the limb having power absorption behavior. They believe that the interaction between muscle powers in able-bodied gait could reflect specific propulsion and control strategies related to each limb (Sadeghi et al., 1997; Sadeghi et al., 2000).

The concept of limb dominance (laterality or limb preference) and its possible relation to symmetrical or asymmetrical behavior of the lower limbs has recently become an issue of interest in biomechanics, but has a longer history of consideration in the motor control and osteopathic arenas. Limb dominance is related to the notion that the two hemispheres of the human brain are functionally dissimilar (Pope, 2003; Porac

& Coren, 1981). Limb preference and laterality are also used to express the preferential use of one limb in voluntary motor acts (Pope, 2003; Porac & Coren, 1981; Sadeghi et al., 2000). Porac and Cohen (1977) defined lower limb laterality by defining “foot preference” measures. The most common measure of self-reported foot/leg dominance is the question of which foot one uses to kick a ball. Other ways suggested to assess footedness/leg dominance are observing the foot and leg that bears the weight during relaxed standing, the foot placed first onto a step during stair ascent, or the leg used for one-legged hopping or jumping maneuvers. Many believe that the foot used in activities is the preferred foot while the non-preferred foot provides postural and stabilizing support (Pope, 2003; Sadeghi et al., 2000). In the literature, mobilization and stabilization are also used to characterize the dominant and non-dominant limbs, respectively. The mobilizing or manipulating limb is the preferred or dominant limb, while the limb used to support or stabilize the action of the preferred limb is the non-preferred limb. Several authors support the general contention that humans are generally right foot dominant for actions of mobilization and left side dominant for postural stabilization and control (Gentry & Gabbard, 1995; Peters, 1988; Sadeghi et al., 2000). Pope (2003), in his discussion of the “Common Compensatory Pattern” of postural asymmetry, reinforces the concepts of cerebral lateralization and vestibular lateralization. The osteopathic theory of cerebral lateralization acknowledges a genetic and developmental basis of left hemispheric dominance; hence right hand and right foot dominance. Osteopathic clinical evidence indicates that fetal growth patterns, labor and delivery patterns and postural control mechanisms after birth direct the presentation of this compensatory pattern (Pope, 2003). Other existing models of hemispheric specialization that relate to limb preference include the *right-shift hypothesis* (Gabbard & Iteya, 1996) that gives additional consideration to environmental factor influences; for those individuals who lack a strong right-shift factor, limb preference is determined at random with the influence of environmental factors. Complementing this line of reasoning is the *right-side world hypothesis* (Porac & Coren, 1981): e.g. machines, desks, doors, toys, and sport items are designed primarily for the right-hander, suggesting that context may influence the magnitude or strength of laterality. In conclusion, it is hypothesized that with general activities of daily living, one leg (left) is primarily used for postural support (vestibular dominance) and the other (right) for most voluntary motor activities (motor dominance) (Pope, 2003).

The influence of laterality has been minimally addressed during gait; to date there is no conclusion regarding the effects on asymmetry profiles in able-bodied gait. Hamill et al. (1984) tested the influence of lateral dominance after reporting the presence of gait symmetry. When ground reaction forces were compared, they did not find any significant differences between the dominant and non-dominant limbs. Gundersen et al. (1989) claimed that there was no relation between foot dominance and gait asymmetry in a gait study.

In contrast, several studies suggested that asymmetrical behavior of the lower limbs was affected by laterality. Singh (1970) claimed that one lower limb does appear as a dominant limb during gait. Matsusaka et al. (1985) reported that medio-lateral balance in walking was controlled by the left, non-dominant limb. Finally, Sadeghi et al. (1997) stated that functional asymmetry in the gait of able-bodied subjects might be related to limb dominance. Although laterality might be considered one explanation of functional gait asymmetry in normal gait, the question has not been settled. However, clinicians such as Hruska (1998) and the Postural Restoration Institute believe that all individuals present with some degree of functional asymmetry which may predispose them to pelvifemoral instability issues and hence, PFP, to name one of many possible orthopedic dysfunctions.

Orthopedic physicians and therapist from the International Patellofemoral Study Group have noted that the term malalignment is frequently used in the description and classification of patellofemoral joint problems. However, many people can function normally even if they present with significant malalignment. Physicians have begun to agree that rather than attempt to define normal and abnormal alignment measures, it may be more useful to determine statistical thresholds for measures of “excess” above which patients begin to have associated problems, such as PFP. This may also be pertinent to research on functional gait asymmetry. We are aware that even able-bodied subjects have high levels of variability on biomechanical parameters of gait. Researchers may need to determine at what level of variability subjects fall out of the range of what is safe to avoid overuse injuries. Zifchock, Davis and Hamill (2006) suggested a 10 point difference in the SI to be clinically relevant in their study. Significantly more research needs to be completed to establish clinically relevant ranges suggestive of injurious consequences.

In conclusion, an individual’s mechanics are dictated, in part, by his or her anatomical structure and alignment. Because aberrant mechanics are thought to lead to injury, a number of attempts have been made to predict injury on the basis of a runner’s anatomical and biomechanical structure. Unfortunately, variability and asymmetry appear to be common to all subjects. Efforts to define more accurately the criteria for abnormal structure and asymmetry are necessary.

CHAPTER III

Methods

Subjects

Ninety-two female athletes ranging in age from 18 to 64 years of age originally volunteered to participate in the study. Forty-six subjects were determined to fit into the patellofemoral pain group (PFP), while an additional forty-six subjects were free of any knee pain or pathology and were assigned to the control, healthy knee group (CHK). Participants were recruited through the Minnesota Distance Running Association, the Twin Cities Marathon Running groups and newsletter and the "Minnesota Dead Runners Society" web site. Contact was made through newsletter and email notices. Additional subjects were recruited from the University of Minnesota general student and staff populations. Four subjects were excluded from the final data analysis. Two PFP subjects fell well outside of the range of ages of the control group (57 years and 64 years) and thus were excluded. Corrected age ranges of both groups after exclusion of these subjects were 18-45 years. The mean age of the PFP group was 29.34 (SD = 7.90); the mean age of the control group was 25.27 (SD = 7.16). The PFP group was significantly older than the control group. Additionally, one PFP subject had recently been diagnosed with hypothyroidism and was excluded secondary to a recent decrease in physical activity level and associated weight gain. One CHK subject was excluded secondary to an incomplete data set. The final sample size was N=88 (PFP = 43, CHK = 45).

Participants were assigned to the PFP group if they met the following inclusion criteria:

1. Female athletes who participated at a moderate recreational or competitive level of athletic participation. Minimum athletic participation requirement was 3 days per week.
2. Ages 18 – 45 years old.
3. Anterior or retropatellar knee pain for more than 4 weeks.
4. Lateral knee pain for more than 4 weeks.
5. Insidious onset of symptoms unrelated to a traumatic event.
6. Pain from at least two of the following activities commonly associated with PFP: prolonged sitting, ascending or descending stairs, rise from sit, squatting, kneeling, running, hopping, or jumping (Crossley et al., 2002; Schneider, Labs, & Wagner, 2001; Timm, 1998).

Individuals with PFP were excluded from this study if they reported one or more of the following exclusion criteria:

1. Pain to palpation along the quadriceps tendon or patellar ligament.
2. Medial plica snapping sensation.
3. Signs and symptoms of meniscal or articular cartilage pathology.
4. Knee joint effusion.
5. History of patellar dislocation.
6. History of a previous knee surgery.
7. Previous use of a Protonics® neuromuscular education system.

The control group subjects were selected based on the same criteria used for the PFP group except these individuals had:

1. No history or diagnosis of knee trauma or pathology.

2. No prior history of ancillary treatment for PFP.

3. No pain with the activities described for the PFP group.

4. No limitations or neurological involvement that would influence gait.

Demographic and health history information was collected from each subject utilizing a subject information questionnaire (Appendix A). Demographic information included age, height (m), weight (kg), Body Mass Index (BMI), hand dominance, leg dominance, side of most pain, knee injury history, duration of symptoms, miles run per week, and level of physical activity based on the number of hours participating in her particular activity per week. The level of subjective lower extremity function was assessed by the Kujala functional knee assessment survey. The Kujala scale was chosen based on its use in previous trials (Arendt, 2000; Crossley et al., 2001; Timm, 1998) and based on its relevance to PFP studies (Appendix B). Subject/group information is listed in Table 1 and Table 2.

Table 1

Subject Demographics based on Group Membership

Subject Characteristic	PFP (n=43)	CHK (n=45)	p-value
Age (years)	29.23 ± 7.90	25.27 ± 7.16	.007*
Height (meters)	1.67 ± .06	1.65 ± .07	.141
Weight (kilograms)	65.85 ± 10.70	63.03 ± 9.23	.178
BMI (kg/m^2)	23.47 ± 3.14	23.02 ± 2.82	.576
Duration symptoms (weeks)	143.21 ± 239.98	0 ± 0	< .001*
Activity hours/week	7.38 ± 4.31	8.34 ± 5.48	.428
Mileage run/week	12.52 ± 13.84	17.40 ± 18.10	.147
Kujala[a]	79.44 ± 9.61	99.38 ± 1.86	< .001*

Note. BMI = Body Mass Index.

[a]Kujala score of 100 indicates no pain or dysfunction.

*p < .05, based on Mann-Whitney U Test.

Table 2

Handedness, Leg Dominance and Side of Pain

Variable	PFP (n=43)	CHK (n=45)	p-value
Handedness			
Right	40	44	0.577
Left	3	1	
Leg Dominance			
Right	30	31	1.000
Left	13	14	
Side of Pain			
Right	16	N/A	
Left	8	N/A	
Bilateral	19	N/A	
Leg Dominance* Side of Pain			0.951

Note. N/A means pain not applicable in control subjects.

Chi-Square, $p < .05$.

Instrumentation

Kinematics. Motion analysis was performed utilizing an Emotion Smart Capture motion analysis system (BTS Bioengineering, Padova, Italy) with 9 cameras operating at a sampling frequency of 60 HZ. Prior to each data collection session, each camera was calibrated and the average spatial resolution was reported to be less than 0.5 mm within an area of 4 m x 3 m x 2 m. Emotion Smart Tracker software was utilized for 3D transformation and path ID modeling of the data points. Three dimensional joint angles were calculated utilizing a customized Matlab program suggested by the University of Calgary, and based on International Society of Biomechanics (ISB) recommendations on definitions of joint coordinate systems (JCS) for the hip and pelvis (Figure 5) (Wu et al., 2002).

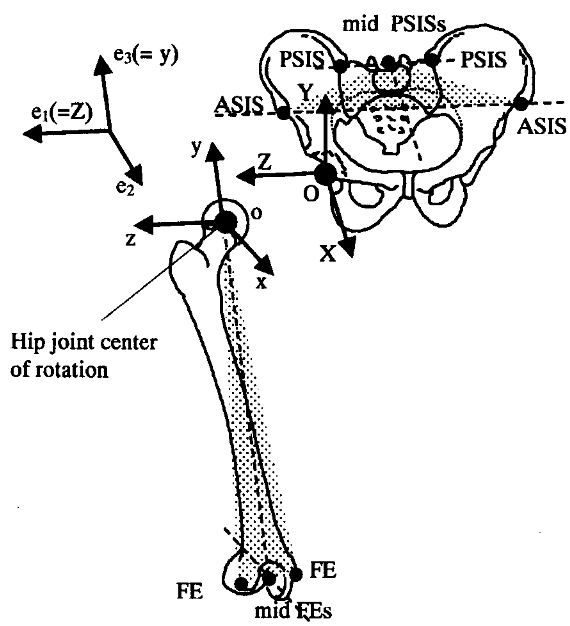


Figure 5. Illustration of the pelvic coordinate system (XYZ), femoral coordinate system (xyz), and the JCS for the right hip joint.

From "ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion-part I: ankle, hip, and spine," by Wu et al, 2002, *Journal of Biomechanics*, *35*, p. 546.

Joint angles were calculated as ordered rotations between anatomically aligned reference frames associated with adjacent body segments (Schutte et al., 2000). The traditionally defined global coordinate system follows the right hand rule and has the positive x-direction oriented in the direction of forward walking or running progression, the positive y-direction oriented upward and the positive z-direction oriented laterally

(Schutte et al., 2000). The standard 3-D angular rotations of the pelvis and hips are computed using a technique equivalent to the geometrical conventions described by Grood and Suntay (1983). The ISB recommends this geometrical convention as a standard for 3-D description of pelvis and hip joint movement (Schache et al., 2001; Wu et al., 2002). ISB recommended pelvic motion is typically measured as rotation of the pelvic segment with respect to the global coordinate system and is isolated to one movement, pelvic tilt, without differentiation of right and left hemi-pelvic motion. Hip motion is measured as rotation of the thigh segment with respect to that pelvic segment (Schache, Blanch, & Murphy, 2000; Wu et al., 2002). Due to frequent movement and set-up/take down of the motion capture system, a global coordinate system was not obtained. In addition, in order to measure kinematic asymmetry of the pelvis, a modified pelvic coordinate system was developed to differentiate right versus left hemi-pelvic motion. The pelvic coordinate system was computed utilizing a modified technique which continued to follow the right-hand rule. The individual right and left pelvic coordinate systems were computed utilizing individualized resting standing frames recorded preliminary to each subject's trial. Subjects stood in their normal resting state and no adjustments to pelvic position were made. Hip motion was measured as rotation of the thigh segment with respect to the standing hemi pelvic segment on both the right and left sides. (See details of pelvis and hip axes calculations in Appendix C.) All data were filtered using a low-pass fourth-order Butterworth filter with a cutoff frequency of 6 Hz. Passive reflective markers were attached to the following boney landmarks based on ISB recommendations (Wu et al., 2002): bilateral anterior superior iliac spines (ASIS), bilateral posterior superior iliac spines (PSIS), a sacral point midway between the two PSIS, bilateral greater trochanters, bilateral medial and lateral femoral epicondyles, bilateral tibial tubercles, bilateral medial and lateral tibial condyles, and the tips of the medial and lateral malleoli. These markers were used to define the embedded coordinate systems of the pelvis, hip, and thigh segments as defined in Figure 5.

Kinetics. Pressure and force data was gathered utilizing a Pedar in-shoe pressure measuring system (©novel electronics inc., St. Paul, MN). The validity and reliability of the Pedar system has been previously documented and has been referenced in Chapter 2. Each pair of insoles used in this study was calibrated prior to the start of data collection utilizing novel's® rubber bladder calibration device. The calibration procedure consisted of applying a linear range of pressures to each insole throughout a measurement range (0-600 kPa). The Pedar components included the insoles, insole cables, and the sync box with connectors for the insole cables and the Pedar cable. There were 99 sensors per insole that provided total foot contact during data collection. Data was gathered at a sampling frequency of 50 HZ. The Pedar box was tethered to a PC computer via a large USB cable. The Pedar acquisition software was used to analyze the raw data.

*Variables measure*d. Primary dependent kinematic and kinetic variables included maximum anterior hemi-pelvic tilt (measured relative to each individual's resting standing pelvic tilt), maximum hip extension, foot contact time, maximum vertical ground reaction force, and a measure of the mediolateral variability of the Center of Pressure (COP) or gait line which we termed the "X-Factor" (COP_x).

Each variable was measured during both walking and running conditions (30 second trials) and for both the right and the left legs. Pelvis and hip ROM measures were recorded as the average maximum degrees of motion obtained throughout ten gait cycles. Contact time was measured as the average time (ms) each foot was in contact with the ground during the 30-second trial. The maximum force measure was the average maximum vertical ground reaction force calculated as a percentage of body weight per limb during the respective 30-second trial. The variable used to describe the change in the COP path, the "X-Factor", was a measure of the standard deviation associated with the average mediolateral shift of the COP path during the recorded movement. This measure was calculated using the standard deviation of the COP measurement along the x axis of the insole in three predetermined foot regions (Figure 6).

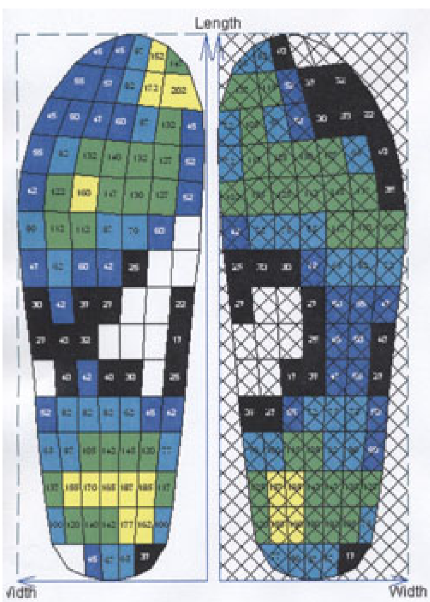


Figure 6. X-Y coordinate system for COP calculations for Pedar insoles. Reprinted with permission, Novel, Inc. St. Paul, MN.

Measures similar to this have become a popular method of describing both normal and abnormal foot movement during gait (Cornwall & McPoil, 2000; Han et al., 1999; Katoh et al., 1983; Nigg et al., 2003). The X-Factor/COP_x variable was calculated utilizing the raw .fgt data files from the Pedar data acquisition program. A Matlab program similar to novel's multimask software program was utilized to divide the foot/insole into three regions: heel, midfoot and forefoot. These regions were based on a percentage of the total foot length and width and were consistently applied to each insole size. The heel mask was measured from 0% to 30% of foot/insole length. The midfoot was measured from 30% to 60% and the forefoot was from 60% to 100% of the foot/insole length. These mask measures have been consistently utilized in the gait literature (Ford et al., 2006; Grampp, Willson, & Kernozek, 2000; VanZant et al., 2001). For each foot, the "Total X-Factor" variable utilized in this study was the total of the "X-Factor" heel + "X-Factor" midfoot + "X-Factor" forefoot. Each of the kinematic and kinetic variables was then expressed as a Symmetry Index (SI) in order to examine the symmetry between the right and left lower limbs during gait. The SI proposed by Robinson et al. (1987) has been widely used in the gait literature. The following equation has been proposed for analyzing symmetry during walking and running (Herzog et al., 1989; Karamanidis, Arampatzis, & Bruggemann, 2003; Robinson et al., 1987):

$$\text{Symmetry Index (SI)} = \frac{(X_R - X_L)}{\frac{1}{2}(X_R + X_L)} \times 100\%,$$

where X_R is the parameter recorded from the right leg and X_L is the corresponding parameter from the left leg. When the SI = 0, there is no difference between the variables and, therefore, there is perfect gait symmetry as measured by that variable. A positive value for Si indicates that the magnitude of the variable on the right is larger than that on the left; a negative value indicates that the magnitude of the variable on the right is smaller than that on the left. SI is measured in % units.

Procedures

Data for each subject was collected during one testing session that lasted approximately 60 minutes. All testing was performed in the Gait and Posture Laboratory, School of Kinesiology, at the University of Minnesota. Prior to participation, informed consent was obtained from each subject in compliance with the University of Minnesota Institutional Review Board (IRB) (Appendix D & E).

Each subject met with the primary investigator (PI) to complete the demographic and health history portion of the subject information form. In addition, each subject took part in one clinical assessment of the pelvis, hips, and femoral alignment. All physical assessments were performed by the PI who is a licensed physical therapist and certified athletic trainer. She was not blinded to group membership as part of her responsibility was establishing if the subject was eligible to participate and in which group she belonged.

The clinical examination included pelvic alignment/symmetry tests, such as the Modified Thomas test (Extension Drop Test), the Ober test (Adduction Drop test), the Adductor Lift test, straight leg raise (SLR), and hip internal and external rotation range of motion (passive and active). The Extension Drop Test, Adduction Drop Test and the

SLR were utilized during data analysis, as these were deemed the most indicative of pelvis symmetry from a clinical perspective.

The Extension Drop test (Appendix F) measured the subject's ability to extend her hip. The subject was positioned supine on a plinth with her knees and hips flexed and the hips close to the edge of the table. With the PI standing in front of the subject, the subject's knees were flexed to the chest until her spine was in a neutral position (flat back). Each leg was passively lowered over the edge of the table while maintaining neutral abduction/adduction of that leg. A neutral spine position was maintained by stabilizing the opposite leg in full hip flexion. This was repeated on both sides. A positive test was indicated by insufficient hip extension (the thigh could not rest on the table) or presence of a snap or pop in the hip joint, indicative of capsular or ligamentous laxity. The outcome measure for this test was either a negative result (documented as a "0" for statistical purposes) or a positive result (documented as a "1".)

The Adduction Drop test (Appendix G) measured the subject's ability to adduct the thigh. The subject was positioned side lying with 70-90 degrees of hip and knee flexion. With the PI standing behind the subject, the subject's top knee was passively flexed to 90 degrees and maintained while abducting and extending the hip to neutral hip extension. The subject's pelvis was stabilized from falling backwards. The tibia was positioned parallel to the tabletop. The femur was not allowed to internally rotate. Passive hip adduction was slowly performed to the subject's top hip and thigh. This was repeated on both sides. A positive test was indicative of insufficient passive adduction. (The medial side of the knee should touch the heel of the bottom extremity if the pelvis is neutral). Positive or negative results were documented as for the Extension Drop test.

The Extension Drop Test and Adduction Drop Test are trademarks of Ron Hruska, PT and the Postural Restoration Institute®, Lincoln, Nebraska. Written permission was obtained to utilize the tests during the project, as well as permission to use pictures and written documentation for this dissertation.

Hamstring length was determined by measuring the straight leg raise (SLR). The SLR test was performed in the supine position with both legs in the extended position. Hamstring length was measured, with a goniometer, as the angle between the horizontal to the tabletop and the femur as the leg was raised with the knee straight. The lower extremity was passively lifted to the end range of motion (firm end feel) or until the examiner noted flexion of the knee or a change in the normal lumbar curve. Hsieh et al. (1983) reported an ICC for intertester reliability of 0.99 for a SLR test using this technique.

Clinical Outcome Measures: Hruska (1998) and the Postural Restoration Institute (PRI) (Hruska & Joutras, 1999) suggest that many patients with patellofemoral pain present with an anteriorly tipped and rotated pelvis. Anterior rotation of the pelvis is also associated with internal rotation of the femur which can result in patellofemoral malalignment, increased lateral patellar pressure and associated hip muscular dysfunction and tone (Hruska, 1998; McCrory et al., 2004; McCrory, Quick, Shapiro, Ballantyne, & Davis, 2007; Powers, 2003; Tyler et al., 2006). PRI suggests utilization of the Extension Drop Test and the Adduction Drop Test as an indicator of pelvic symmetry. Based on the results of the Extension and Adduction Drop tests, two indices were established. Since a positive Extension Drop test is believed to indicate over activity or increased functional tone in the hip flexors and a positive Adduction Drop

Test is believed to indicate a forward position of the iliac or ilea and associated lateral hip tightness, a "Tightness Index" (TI) was established. The TI was based on the total number of positive tests as determined by bilateral Extension Drop and Adduction Drop test results. The TI variable was recorded on a 0 – 4 scale with 0 (four negative tests) equivalent to normal tightness and 4 (four positive tests) equating to a maximum tightness level (Appendix H). This index is thought to be more indicative of a more traditional physical therapy approach to assessment of "muscle tightness". A second index, the Pelvic Symmetry Index (PSI), related results to a level of pelvic asymmetry that was specific to right vs. left side involvement (Appendix G). PRI and others (Hruska & Joutras, 1999; Pope, 2003; Porac & Coren, 1981) suggest that based on dominance factors, environmental influences, labor and delivery stresses and brain lateralization patterns that the majority of subjects establish weakness and pelvic instability patterns on the left side; the right side is predominantly stronger and more stable in unilateral stance activities. To evaluate these suggestions, a PSI was established based on the number and side of positive pelvic symmetry tests. The PSI variable was also recorded based on a 0 – 3 scale (0 = four negative tests, 3 = 4 positive tests, and 1 an 2 equating to increasing levels of positive tests on each of the left and right sides). The SLR measure was also expressed as a Symmetry Index for analysis purposes.

During the kinematic and kinetic data collection, each subject was asked to wear spandex running tights/shorts, a jog bra, and short cotton socks. Passive reflective markers were positioned over specific anatomical boney landmarks on the pelvis, hips, femur, tibia, and ankles. The PI performed all marker placements to avoid inter-tester variability.

To permit standardization of footwear used during testing, the Pedar insoles were fitted into Aqua socks (Target® brand) that had a cloth upper and rubber outsole. Aqua socks were utilized to negate the effects of orthotics, arch support, and motion control measures on lower extremity alignment further up the lower extremity kinetic chain. Each subject wore cotton socks in the aqua sock shoes. Pedar insole cables were secured to the subject's lower legs utilizing saran wrap. The Pedar data gathering unit was held by an assistant in an effort to minimize the effect of the weight of the Pedar box on walking and running performance. Kinematic and kinetic data were collected simultaneously with the Pedar system synchronized manually (with verbal cues) to the optoelectric camera system.

All walking and running trials were conducted on a large treadmill. The participants were asked to walk and run at a self-selected walking and running speed. Each participant was asked to walk at a "brisk" pace and to run at a speed they could tolerate for at least 5 minutes. Each subject was given adequate time to adjust to the speed of the treadmill. Once the subject achieved her self-selected speed, both kinematic and kinetic data were collected. Each subject performed two trials of 30 seconds each of walking and running. The subject utilized the same speed for each of the two trials, respectively.

Data Analysis

Statistical analyses were performed using SPSS Statistical Analysis software for Windows, version 14.0 (SPSS Inc., Chicago, IL, USA). A priori calculations for sample size were not performed; sample size estimation was based off of the minimal recommendations of 4 to 5 subjects per variable analyzed in a multivariate analysis statistic (Hair, Anderson, Tatham, & Black, 1998). All demographic, clinical, kinematic and kinetic parameters were assessed for normality utilizing a Shapiro-Wilk statistic and graphical methods prior to statistical analysis. A number of demographic and SI measures violated one or more of the assumptions of normality. In those individual cases non-parametric statistics were performed. Data transformation was not performed on the variables that failed the normality assessment as negative SI measures were pertinent to the statistical output; transformations are not recommended if negative values are part of the parameter output.

Independent-samples t-tests and the Mann-Whitney U test were used to check for differences in demographic information between groups. An alpha of .05 was used in these comparisons. The significance of the difference in clinical, kinematic and kinetic presentation between groups was also determined utilizing independent-samples t-tests and Mann-Whitney U tests for all continuous variables and Pearson’s Chi-Square test for the categorical and ranked data. In addition, logistic regression analysis was performed on the specific index categories of the TI and PSI as pairwise comparisons between groups. An alpha of .05 was also used in these comparisons.

Logistic regression was then applied to the clinical, kinematic and kinetic measures to determine which measures discriminate between the groups with and without patellofemoral pain. Conditional logistic regression was chosen as it is recommended for case-controlled analysis methods. Logistic regression analysis makes no assumptions about the distributions of the predictor variables and can be used with any mix of continuous, discrete and dichotomous variables and was, thus, the most appropriate statistical analysis for this data set. The dichotomous grouping variable (independent variable) was patellofemoral pain versus no pain. The predictor variables were:

- Age
- Tightness Index
- Pelvic Symmetry Index
- Straight leg raise
- Contact time
- Maximum vertical ground reaction force
- "X-Factor"/ COP_x
- Anterior pelvic tilt ROM
- Hip extension ROM

The set of predictors which most reliably distinguished between the groups was selected and reported.

CHAPTER IV

Results

Subjects

Subject characteristics of both groups are reported in Table 1. A significant difference was found for the mean age ($p = .007$) between the groups, with the PFP group being older. As expected, the two groups were also significantly different in terms of duration of pain ($p < .001$) and Kujala score ($p < .001$). The Kujala score (M= 79.44, SD= 9.61) indicates a mild to moderate level of disability within the PFP group. There were no significant differences in height, weight, BMI, level of activity, and miles run/week. An alpha level of .05 was used for all statistical analyses.

Table 2 summarizes handedness, leg dominance and side of pain results. There were no significant differences between handedness and leg dominance between groups; leg dominance percentages were nearly identical between groups (CHK: left, 31.1%, right, 68.9%; PFP: left, 30.2%, right, 69.8%). There was also no significant pattern of leg dominance versus painful side in the PFP group, X^2 (2, N=43) = 0.10, p=.951 (Figure 7). In other words, subjects in the PFP group did not experience more knee pain in their dominant leg more often than in their non-dominant leg.

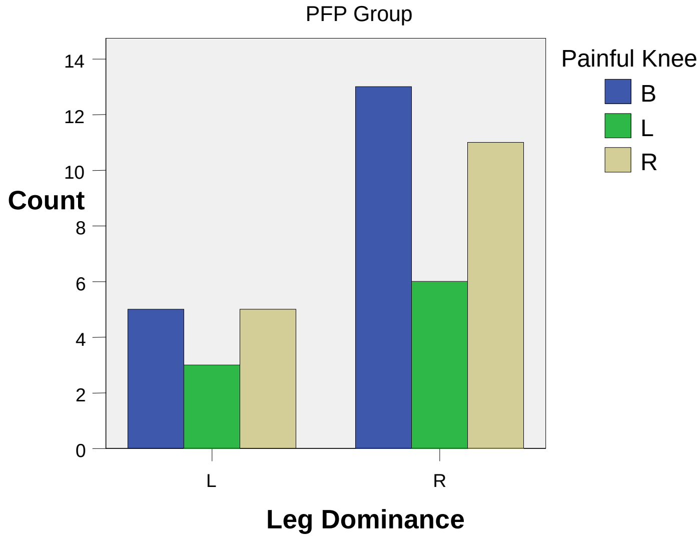


Figure 7. Side of knee pain versus leg dominance in PFP group.

Note: B = bilateral knees, L = left knee, and R = right knee.

Clinical Outcome Measures

To test hypothesis 1, we used Pearson's Chi-square analysis for this categorical data. The results of the Chi-square analysis for left and right Extension Drop/Thomas tests and Adduction Drop/Ober tests are indicated in Table 3. There are statistically significant differences between the PFP and CHK groups on all 4 tests; the proportion of PFP subjects with positive results is significantly greater than CHK subjects. The odds ratio for each clinical test is presented in Table 4.

Table 3

Prevalence (%) of Left and Right Extension Drop/Thomas and Adduction Drop/Ober Tests

	PFP		CHK			
	(+)	(-)	(+)	(-)	X^2 (1)	p-values
Thomas (L)	46.5	53.5	15.6	84.4	8.51	.004*
Thomas (R)	25.6	74.4	6.7	93.3	4.55	.033*
Ober (L)	83.7	16.3	46.7	53.3	11.66	.001*
Ober (R)	60.5	39.5	22.2	77.8	11.77	.001*

Note: Test results measured as “positive” or “negative”.

*p < .05

Table 4

Odds Ratios for Left and Right Extension Drop/Thomas and Adduction Drop/Ober Tests

	PFP		CHK		
	(+)	(-)	(+)	(-)	Odds Ratio
Thomas (L)	20	23	7	38	4.83
Thomas (R)	11	32	3	42	4.84
Ober (L)	36	7	21	24	5.87
Ober (R)	26	17	10	35	5.35

Note: Odds ratio is calculated based on frequency of positive and negative results.

Based on the odds ratios, the PFP subjects were 4.83 times more likely to present with a positive left Thomas test, 4.84 times more likely to present with a positive right Thomas test, 5.87 times more likely to present with a positive left Ober test, and 5.35 times more likely to present with a positive right Ober test than the CHK subjects.

The clinical outcomes were also assessed utilizing two defined categorical indices. Results of Chi-square analyses of the “Tightness Index” (TI) and “Pelvic Symmetry Index” (PSI) are summarized in Tables 5 and 7, respectively. Figures 8 and 9 represent graphical breakdown of TI and PSI based on group membership. Results of Logistic Regression predicting group membership with the TI and PSI are presented in Tables 6 and 8, respectively.

Table 5

Chi-square Results of Tightness Index Versus Group Membership

	Tightness Index				
Group	4 (-)	1 (+)	2 (+)	3 (+)	4 (+)
PFP (%)	11.6	14.0	39.5	11.6	23.3
CHK (%)	51.1	22.2	17.8	2.2	6.7

Note: Tightness index based on the total number of positive or negative results of left and right Thomas and Ober tests.

There was a significant group difference in the Tightness Index, X^2 (4, N = 88) = 22.21, p < .001, Φ_c = < 0.001. Logistic regression analysis of the specific index levels indicated that there was no significant difference between groups when “one positive test” was measured (p = 0.155); in fact, the CHK group presented with a greater frequency of “one positive” test than the PFP group. However, all remaining levels of the index significantly differentiated the PFP group clinically from the CHK group. Control group participants were 4.39 times more likely to present with “normal” tightness (“four negative tests”) as compared to PFP subjects. The logistic regression model for the TI was able to correctly classify 73.3% of the CHK group and 74.4% of the PFP group, for an overall success rate of 73.9%.

Table 6

Logistic Regression Predicting Group Membership with the Tightness Index

Predictor	B	Wald X^2	p	Odds Ratio	95% CI
Tightness Index		19.14	.001		
One (+) test	1.015	2.02	.155	2.76	0.68 to 11.19
Two (+) tests	2.28	12.16	<.001	9.78	2.72 to 35.2
Three (+) tests	3.14	6.81	.009	23.00	2.18 to 242.33
Four (+) tests	2.73	11.01	.001	15.33	3.10 to 76.90

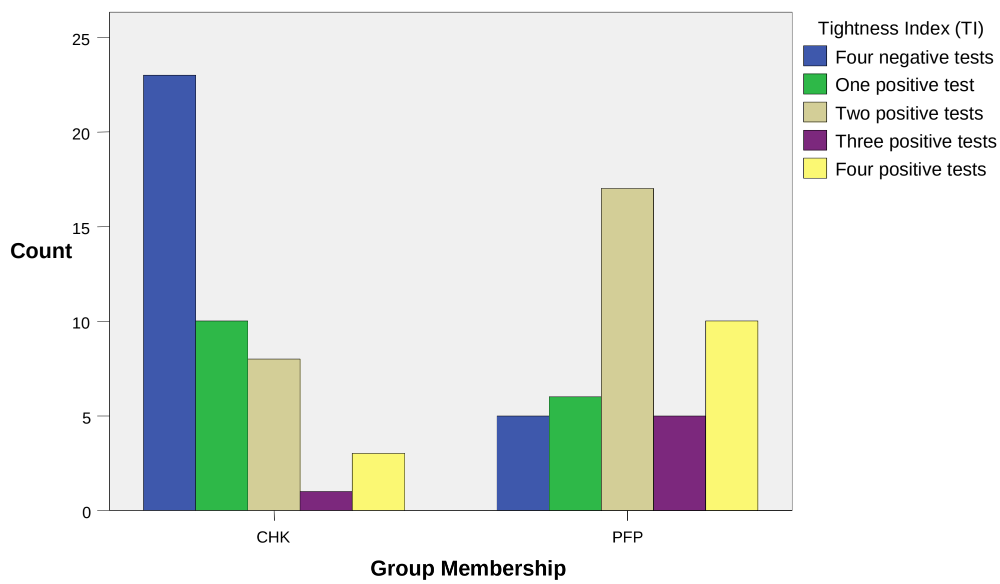


Figure 8. Tightness index as a predictor of group membership.

Table 7

Chi-square results of Pelvic Symmetry Index versus group membership

Group	4 (-)	Pelvic Symmetry Index (+) L Ober and/or L Thomas	(+) L & R Ober or Thomas	4 (+) Ober & Thomas
PFP (%)	11.6	23.3	41.9	23.3
CHK (%)	51.1	28.9	13.3	6.7

Note: Pelvic symmetry index based on specific representation(s) of positive left and right Thomas and Ober tests.

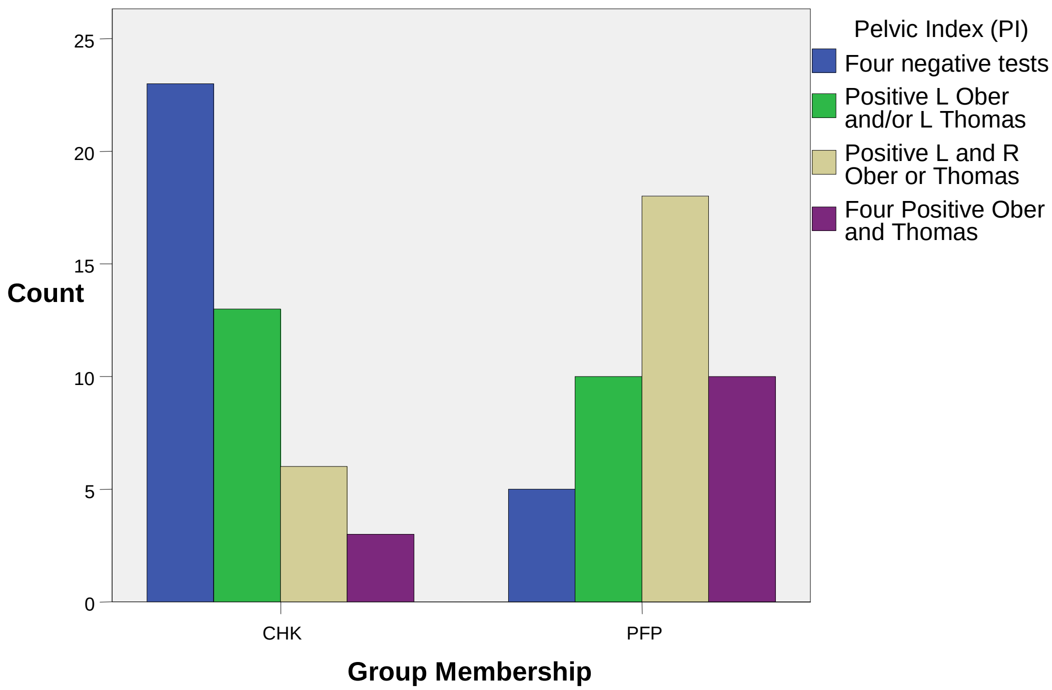


Figure 9. Pelvic symmetry index as a predictor of group membership.

There was an additional significant relationship between group membership and the Pelvic Symmetry Index, X^2 (3, N = 88) = 21.70, p < .001, Φ_c = < 0.001. Logistic regression analysis of the specific index levels indicated that there were significant group differences within this index. PFP subjects were significantly more prone to exhibit bilateral presentations of clinical tests and four positive tests. The CHK group was 4.25 times more likely to present with all negative tests and were surprisingly more likely to present with left sided positive tests only. The logistic regression model was able to correctly classify 80.0% of the CHK group and 65.1% of the PFP group, for an overall success rate of 72.7%.

Table 8

Logistic Regression Predicting Group Membership with the Pelvic Symmetry Index

Predictor	B	Wald X^2	p	Odds Ratio	95% CI
Pelvic Symmetry Index		18.73	<.001		
+ L Ober ± L Thomas	1.26	3.80	.051	3.54	0.99 to12.61
+ L & R Ober or Thomas	2.63	14.79	<.001	13.80	3.62 to 52.57
Four (+) tests	2.73	11.01	.001	15.33	3.06 to 76.90

An independent-samples t-test was conducted to compare the straight leg raise (SLR) symmetry index for PFP and CHK. There was no significant difference in SI SLR for CHK (M=1.21, SD=7.65) and PFP (M= -.38, SD=11.15; t (86) = -.780, p=.438). The magnitude of the differences in the means was very small (eta squared= .007).

Biomechanical Outcome Measures

To test hypotheses 2, 3, & 4, univariate data analysis was performed to assess group differences for each biomechanical variable in both conditions. Means and SD of the raw data (kinematic and kinetic variables) are presented in Appendix I and J.

Walk. The independent-samples t-test and Mann-Whitney U test showed no significant difference between groups on the symmetry indices for pelvic tilt ROM, hip extension ROM, maximum vertical GRF (% body weight) and contact time, respectively. The SI for the "X Factor" or mediolateral shift of COP approached significance ($p = 0.055$) with the PFP group exhibiting greater variability of the gait line on the left side as compared to the CHK group. Figure 10 illustrates exemplar mediolateral COP shift data in representative examples for both PFP and control subjects. As was seen with the clinical data, there were incidences of both stable and unstable COP for both groups. In general, however, PFP subjects showed greater mediolateral COP variability or shift on the left vs. the right foot. Table 9 shows means, standard deviations, and t-test results of symmetry indices for all kinematic and kinetic parameters during walking.

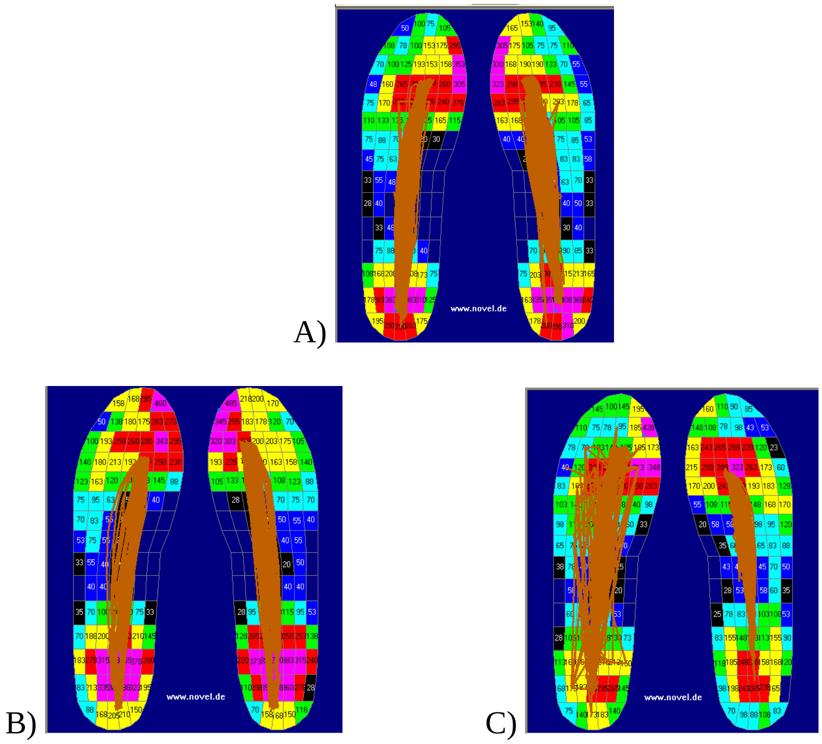


Figure 10. COP/gait line results for representative PFP and control subjects during walking. A) Control subject with bilateral stable gait line, B) PFP subject with bilateral gait line stability, and D) PFP with left sided mediolateral gait line variability.

Table 9

Means, SD, and Level of Significance for T-Tests Analyzing Symmetry Indices for Kinematic and Kinetic Variables (Walk)

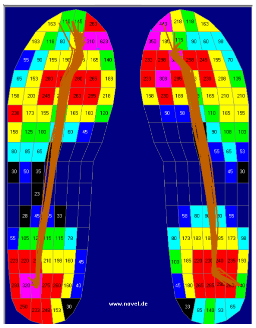
	PFP SI Mean (SD)	CHK SI Mean (SD)	p-value
SLR (deg)	-0.382 (11.15)	1.214 (7.65)	0.438
Pelvic Tilt (deg) ‡	1.77 (36.16)	-0.874 (26.73)	0.846
Hip Extension (deg) ‡	8.77 (53.41)	-4.55 (67.24)	0.618
Maximum force (%BW)	1.008 (10.34)	.884 (7.63)	0.949
Contact time (ms)	3.76 (7.20)	1.76 (10.93)	0.859†
COP$_x$ (mm)	-7.75 (31.69)	4.61 (27.77)	0.055*

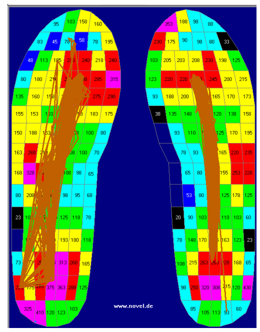
Note. SLR is a clinical kinematic variable. SI = % units.

† Based on Mann-Whitney U Test.

*$p < 0.05$; ‡ Sample size smaller for these variables (n=22).

Run. Table 10 shows results of univariate testing of symmetry indices for kinematic and kinetic parameters during running. Mann-Whitney U test results showed no significant group difference for contact time (p = .735). Independent samples t-tests were conducted to compare the SI of pelvic tilt and hip extension ROM between groups. There was no significant difference in scores between the two subject groups on hip extension SI (p =0.636) or the anterior pelvic tilt SI (p = 0.061). While not statistically significant (p = .085), the X Factor/COP_x again, showed a trend toward more mediolateral shift of the COP/gait line of the left foot of the PFP group. COP_x/gait line comparisons of PFP and CHK subjects during running are illustrated in Figure 11. Based upon results of a Mann-Whitney U test, there was a significant group difference (p = .019) for maximum vertical GRF for the CHK group. The control group presented with a -3% SI, indicating greater maximum vertical GRF generated through the left limb than the right.

A) 

B) 

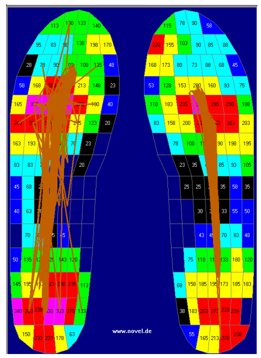
C) 

Figure 11. COP/gait line results for representative PFP and control subjects during running. A) Control subject with significant bilateral gait line stability, B) PFP subject with mild left sided gait line variability, and C) PFP subject with significantly different right vs. left gait line patterns.

Table 10

Means, SD, and Levels of Significance for T-Tests Analyzing Symmetry Indices for Kinematic and Kinetic Variables (Run)

	PFP SI Mean (SD)	CHK SI Mean (SD)	p-value
Pelvic Tilt (deg) ‡	-28.93 (44.97)	17.97 (53.81)	0.061
Hip Extension (deg) ‡	2.3 (19.61)	9.84 (46.44)	0.636
Maximum force (%BW)	.705 (7.52)	-3.01 (11.61)	0.019*†
Contact time (ms)	5.97 (10.89)	11.20 (20.94)	0.735†
COP_x (mm)	-20.73 (43.17)	-4.54 (42.96)	0.085

Note. SLR is a clinical kinematic variable. SI = % units.

† Based on Mann-Whitney U Test.

*$p < 0.05$.

‡ Sample size smaller for these variables (n=22).

Logistic Regression

Logistic regression analysis was performed to determine if a set of clinical and/or biomechanical variables existed that could predict group membership (hypothesis 5). The predictor variables proposed for the full model were age, TI, PSI, SI SLR, SI contact time, SI maximum vertical GRF, SI total X Factor/COP_x, SI pelvic tilt and SI hip extension. However, due to a significant difference in sample size (n=22) between the pelvic and hip ROM variables and the remainder of the biomechanical data, pelvic tilt and hip extension ROM data were not placed into the logistic regression equation. Since logistic regression is sensitive to high correlations among predictor variables, a crosstabs procedure was performed between the TI and the PSI variables. High correlations were found between the variables for both the CHK (r=.945) and the

PFP group (r=.905). Since the TI was found to be more predictive of the PFP group, the PSI variable was deleted from the model to eliminate multicollinearity.

Walk. A test of the full model versus a model with intercept only was statistically significant, χ^2 (9, N = 88) = 33.23, p < .001. The model was able to correctly classify 72.7% of CHK participants and 72.1% of the PFP group, for an overall success rate of 72.4%. The Cox & Snell R^2 was 0.317 and the Nagelkerke R^2 was 0.423, suggesting that between 31.7 and 42.3 percent of the variability is explained by this set of variables.

Table 11 shows the logistic regression coefficients, Wald tests, odds ratios, and 95% confidence intervals for each of the predictors during the walk condition. Employing a .05 level of significance, the TI and 3 of the 4 TI dummy variables had significant partial effects. The variables age and COP_x approached significant levels at p = 0.065 and p = 0.072, respectively.

Table 11

Logistic Regression Analysis Descriptive Statistics (Walk)

Predictor	B	Wald X^2	p	Odds Ratio	95% CI
Age	.070	3.39	.065	1.073	0.996 to 1.16
Tightness Index		17.79	.001*		
One (+)	6.95	.771	.380	2.00	0.43 to 9.44
Two (+)	2.29	10.49	.001*	9.84	2.47 to 39.27
Three (+)	2.95	5.63	.018*	19.14	1.67 to 219.43
Four (+)	2.98	10.54	.001*	19.68	3.26 to 118.95
SI_SLR	-.037	1.50	.221	.964	0.91 to 1.02
SI_Contact time	-.001	.001	.975	.999	0.93 to 1.07
SI_Max Force	-.019	.391	.532	.981	0.93 to 1.04
SI_COP_x	-.019	3.24	.072	.982	0.96 to 1.002
Constant	-3.39	8.11	.004	.034	

*p < .05.

The TI was 5 to 18 times more effective as a predictor of PFP as all other predictor variables during walking. Age approached significance as an effective predictor of group membership as did the Total X Factor/COP_x variable.

Run. Logistic regression analysis utilizing the same predictor variables was also performed under the “run” condition. A test of the full model versus a model with intercept only was statistically significant, χ^2 (9, N = 88) = 40.11, p < .001. The model was able to correctly classify 81.4% of CHK participants and 76.7% of the PFP group, for an overall success rate of 79.1%. The Cox & Snell R^2 was 0.373 and the Nagelkerke

R^2 was 0.497, suggesting that between 37.3 and 49.7 percent of the variability is explained by this set of variables.

Table 12 shows the logistic regression coefficients, Wald tests, odds ratios, and 95% confidence intervals for each of the predictors during running. Employing a .05 level of significance, age, TI and 3 of 4 dummy variables, SI contact time and SI Total X Factor/COP_x had significant partial effects. During running, the most effective predictors of group membership appear to be age, TI, and COP_x.

Table 12

Logistic Regression Analysis Descriptive Statistics (Run)

Predictor	B	Wald X^2	p	Odds Ratio	95% CI
Age	.110	7.53	.006*	1.12	1.032 to 1.21
Tightness Index		16.81	.002*		
One (+)	.732	.777	.378	2.08	0.41 to 10.60
Two (+)	2.73	11.23	.001*	15.40	3.11 to 76.19
Three (+)	2.78	3.77	.052	16.14	0.97 to 267.46
Four (+)	3.35	10.96	.001*	28.54	3.92 to 207.60
SI_SLR	-.025	.668	.414	.975	0.92 to 1.04
SI_Contact time	-.059	5.61	.018*	.943	0.90 to 0.99
SI_Max Force	.006	.041	.840	1.01	0.95 to 1.07
SI_COP_x	-.020	6.22	.013*	.980	0.965 to 0.996
Constant	-4.44	11.59	.001	.001	

*p < .05

The TI was 2 to 29 times more effective as a predictor of PFP as all other predictor variables during running. Age was also a predictor of group membership during this condition.

CHAPTER V

Discussion

The purpose of this study was threefold. First, we attempted to establish a clinical profile of symptomatic female PFP subjects based on assessments recommended by the Postural Restoration Institute®. Second, we attempted to establish a biomechanical profile of these same PFP subjects. Third, we tried to determine if there was a set of biomechanical variables that discriminated between subjects with and without PFP that could be used alone or with our current clinical assessment tools. Within the cohort tested, logistic regression analysis of clinical and biomechanical variables revealed that, in addition to age of the subject, two clinical and one biomechanical variable were predictive variables for PFP. These factors are (1) the Tightness Index (TI); (2) the Pelvic Symmetry Index (PSI); and (3) the Total X Factor/COPx measured during running. The TI and PSI were established based on positive or negative results on the Modified Thomas and Ober tests, with greater frequency of positive results being associated with the PFP group.

Clinical Outcome Measures

While physicians and clinicians have a wide variety of clinical assessment tools to evaluate patients with lower extremity dysfunctions, the Postural Restoration Institute (PRI) has suggested the use of 4 quick and relatively simple screening tools to evaluate patients with suspected pelvifemoral control issues (e.g., PFP). These tests include the Modified Thomas test (Extension Drop test), the Modified Ober (Adduction Drop test), supine trunk rotation, and the lateral step down test (Hruska & Joutras, 1999). They suggest that even one positive result on these tests warrants further evaluation of the pelvifemoral control issue. It is also believed that the greater the number of positive results, the greater the indication of dysfunction (Arendt, 2000; Hruska & Joutras, 1999). Optional assessment tests include the straight leg raise (SLR). Based on these PRI recommendations and assessments of numerous patients, it is believed that the Modified Thomas, Modified Ober and SLR tests are key assessment tools in the evaluation of patients with PFP (Piva et al., 2006; Post, 1999) and in the evaluation of running injuries (Plastaras, Rittenberg, Rittenberg, Press, & Akuthota, 2005).

The results of our primary analysis indicate that subjects with PFP exhibited significantly greater frequency of positive results on two of the three clinical pelvic symmetry tests as stated in hypothesis 1. Those tests were the Modified Thomas/Extension Drop test and the Modified Ober/Adduction Drop test. The results of the Chi-square analysis for left and right Modified Thomas and Modified Ober tests indicates that the proportion of PFP subjects with positive results is significantly greater than CHK subjects on all four measures. Only one other study has looked at the results of these tests based on this PRI approach. Brown et al. (2007) looked at the use of the Modified Thomas and Ober tests as predictors of pelvifemoral malalignment in patients with low back or knee pain. They found that 29 out of 35 subjects had a positive left Ober test compared to 19 out of 35 controls. Within their subject group, a positive left versus right Ober test was also significant. However, they found that no significant difference was observed in the Thomas test. This may have been due to the fact that they had a smaller sample size.

Based on these individual clinical test results, two unique indices were developed, the “Tightness Index” and the “Pelvic Symmetry Index” (Appendix H). The

TI looks at the results from a more traditional physical therapy perspective of muscle and joint tightness. In most cases, patients with increased tightness patterns tend to exhibit more musculoskeletal dysfunction or injury. Data from our study indicated that there was no significant difference between groups when only “one positive test” was assessed; in fact, the control group presented with a greater frequency of “one positive” test than the PFP group. However, as the number of positive results increased the corresponding level of the index significantly differentiated the PFP group from the CHK group. In addition, control subjects were 4 times more likely to present with “normal” tightness (“four negative tests”) as compared to PFP subjects.

The PSI looks at the pelvic symmetry data from an alternative thought process or explanation developed by the Postural Restoration Institute®. The Postural Restoration Institute® has described the Anterior Interior Chain Pattern (AIC) relating to kinetic chain dysfunction affecting back and knee pain (Hruska & Joutras, 1999). This pattern is common to all humans and contributes to postural asymmetry. How people compensate for this pattern can vary, however, the underlying dominant pattern exists in everyone. This “common compensatory pattern” has also been described clearly by Pope (2003). Based on brain laterality research (Hruska & Joutras, 1999; Pope, 2003; Sadeghi et al., 2000), foot dominance patterns (Gabbard & Iteya, 1996; Singh, 1970), internal organ placement and asymmetry, primitive reflexes and vestibular imbalances (Hruska & Joutras, 1999; Pope, 2003), gravity and environmental factors, we tend to stand more on our right leg and compensate with rotation of our trunk and abdominal area to the left. This leads to presentations of greater strength, power and dominance of the right lower extremity. Subjects present with an anteriorly tipped and forwardly rotated pelvis on the left secondary to weakness of the left hamstring, gluteus maximus and left abdominal obliques. Based on this pattern, physical therapists may see an increased number of positive modified Thomas and Ober tests on the left and secondarily (due to increased levels of compensation) on the right. Based on the above pelvic asymmetry, one would also expect changes in SLR presentation (R vs. L).

Our PSI data also showed significant group membership distinctions. PFP subjects were significantly more prone to exhibit bilateral presentations of clinical tests (bilateral L and R Thomas or Ober) and four positive tests (bilateral right and left Thomas and Ober). The CHK group was 4 times more likely to present with all negative tests and were surprisingly more likely to present with left sided positive tests only (positive left Thomas and/or left Ober). While this result may refute the teachings of PRI (i.e., further evaluation of the patient with at least one positive test), it may be that these subjects are more predisposed to future injury based on this pattern. These subjects were significantly younger as a group and did not report symptoms based on our inclusion criteria. More research needs to be done to look at these patterns in pathological and normal subjects.

Finally, we found no difference between the right and left SLR between groups as measured by a Symmetry Index. This study appears to be the only one that has looked at right vs. left leg differences in a SLR assessment. While several studies have looked at hamstring flexibility in subjects with and without PFP, these studies only include the involved limb in comparison with the analogous limb of a control subject. Piva et al. (2005) found that patients with PFP seemed to have less hamstring flexibility compared to individuals without PFP. Their findings were similar to the results of Smith et al. (1991) but contrasted with the data by Witrovouw et al. (2000). Smith and associates (1991) used a cross-sectional design to explore the association between the presence of hamstring tightness and PFP in figure skaters and reported a significant association between these variables. Witrovouw et al. (2000) investigated risk factors in the development of PFP using a longitudinal design and concluded that hamstring flexibility was not a significant factor as assessed by SLR in athletes who developed PFP and those who did not. While Piva and associates (2005) felt the difference in their results compared to Witvrouw et al. (2000) may have been due to an older subject population (20 to 42 years vs. 17 to 21 years), the subjects in our PFP were significantly older than the control group, but exhibited no difference in hamstring symmetry compared to a younger control group. More recently, White and associates (2008) looked for differences in hamstring length between patients with PFP and healthy asymptomatic control participants between 18 and 35 years old. They found that hamstring length was shorter in the PFP group; however, hamstring length was evaluated utilizing the passive knee extension method as measured at the popliteal angle of the knee. The previously noted studies all used the SLR measure of hamstring length. Further research on this subject should be done looking at symmetry between legs in patient population groups as opposed to comparisons of unilateral injured extremities versus control limbs.

Biomechanical Outcome Measures

The second aim of this study was to establish a biomechanical profile of female athletes with PFP. The kinematic and kinetic biomechanical variables were chosen based on their association to the clinical outcomes.

Kinematics- Contact time: In hypothesis 2 we proposed that ground contact time patterns would be different between the two groups. Numerous authors purport that during gait the lower limbs maintain functional roles. This evidence appears frequently in the able-bodied gait literature (Goble et al., 2003; Hirokawa, 1989; Matsusaka et al., 1985; Pope, 2003; Sadeghi et al., 2000). Generally it is accepted that the left or non-dominant limb is used in a supportive manner in gait while the right or dominant limb performs more of a propulsive function. This distinction of roles between limbs is often referred to as a “functional asymmetry” and is thought to be a solution to the two main goals of human locomotion: stability and speed (Goble et al., 2003).

Based on univariate statistical analyses, we found no significant difference between the groups on contact time during either the walk or run conditions. Goble et al. (2003) noted that at slower velocities, able-bodied subjects spent a larger amount of time on the left foot in the stance phase of walking than the right. Having a longer stance time during gait is often associated with an attempt toward greater stability as can be seen in some pathologic gait patterns (e.g., amputee gait). This did not hold true for our PFP subjects. While PRI theory indicates that patients with pelvifemoral dysfunction are weaker and more unstable on the left lower extremity, one might expect these subjects to increase their contact time on the left in an effort to improve stability or, in contrast, to spend more time on the right in an effort to utilize the stronger limb for stability. This was not evident in our sample. However, since our PFP subject group tended to present with more bilateral clinical presentations, the right sided compensation patterns may have affected their gait kinematics during walking. Goble et al. (2003) is also frequently referenced when researchers explore the effect of horizontal velocity changes on kinetic and kinematic gait variables. They proposed a trend toward improved symmetry at higher velocities. However, this is based on data they collected at slow and fast speeds that were only 10% lower or higher than the “normal” self-selected speed chosen by their subjects. However, they proposed that with greater increases in speed it would be more likely that asymmetries could be evoked if they existed. This was not evident within our running data, as there was no significant difference between groups. However, subjects in our study were allowed to establish self-selected walk and run speeds; perhaps many of the subjects could have run faster during the run condition. Our data were similar to Hreljac et al. (2000) who found no difference between right and left limb contact time for a group of runners with one or more overuse running injuries as compared to a group of runners who were injury-free. Thijs et al. (2007) also looked at contact time measures between military recruits who developed PFP during basic training and those who did not. They also found no differences between the two groups; however, data from both legs of the same subject were collapsed together so symmetry between legs could not be assessed. Based on our data, there is no evidence to support hypothesis 2 that ground contact time distribution patterns would be different between our two groups.

Kinematics- Range of motion measures: In addition, we looked at two range of motion measures, anterior pelvic tilt and hip extension. The hip extension ROM was measured as the maximum hip extension range attained during the stance phase of gait; anterior pelvic tilt was measured at the time of terminal hip extension. In hypothesis 3, we stated that there would be group differences in these measures. Schache et al. (2000)

noted that increased tightness or tone in the hip flexor musculature of an athletic population may reduce hip extension flexibility. They proposed that limited hip extension flexibility may be one cause of increased anterior pelvic tilt in runners We can assess this clinically utilizing the Modified Thomas or Extension Drop test. Clinicians believe there is a relationship between anterior pelvic tilt and peak hip extension during gait and hip extension flexibility measured clinically.

Based on our data, there was no significant difference between the PFP and the control groups in measures of anterior pelvic tilt and hip extension ROM during either the walk or the run condition. While the results of the anterior pelvic tilt during running might suggest that there is more movement occurring on the left side of the pelvic innominate in PFP subjects (M= -28.93, SD = 44.97) than in the CHK group (M=17.97, SD=53.81), we must be careful when interpreting the results. Standard deviation measures indicate significant variability within the data and the sample size was smaller. Some of our subjects did not have a kinematic walk or run trial based on our minimal criteria of 10 or more seconds of consistent data. Our results, however, do agree with the findings of Schache et al. (2000). They found no significant difference between the left and right sides for anterior tilt and hip extension ROM in 14 elite track and field athletes. They also found that hip extension flexibility, measured by the Modified Thomas test, was not found to be indicative of the dynamic measures of peak hip extension ROM or anterior pelvic tilt (Schache et al., 2000). Bar-On and associates (1992) reported similar findings in 51 neurological patients with hip flexion contractures. They also found no correlation between Thomas test results and radiologically determined measures of hip extension on the same subjects. Schache and colleagues (2000) suggested possible explanations as to why Thomas test measures were not found to relate to the dynamic measures. "It may be that static flexibility is not the primary factor governing the degree of anterior tilt or peak hip extension ROM when running at a submaximal speed. Such variables may be determined by complex dynamic neuromotor patterns rather than static flexibility alone" (p. 282). It may be that static measures only become a factor at maximal speeds of running when static flexibility is maximized and greater ROM is stressed. Much more research needs to be done in this area at both submaximal and maximal running speeds. Based on our results, there is no evidence to support hypothesis 3 that anterior pelvic tilt and hip extension ROM measures would be different between the groups.

Kinetics-Vertical ground reaction forces: In hypothesis 4, we proposed that significant kinetic differences would be present between groups as measured by maximum ground reaction force (GRF) distribution and center of pressure patterns during both walking and running. Our results showed that there were no differences between groups on maximum GRF symmetry (as a percentage of body weight) during walking, but was statistically significant during the running condition. These results contradicted those of Goble et al. (2003) who found significant differences in the mean left/right differences between feet at slower walking speeds in a normal population. The asymmetry indicated a tendency toward larger force production by the right foot as opposed to the left. This did appear to confirm the use of the right or dominant limb for a propulsive role during gait.

Few studies have looked at symmetry of ground reaction forces in patient populations (e.g., PFP) versus controls. Hreljac et al. (2000) found that there were no significant differences between right and left foot values for any of their biomechanical variables when comparing injured runners versus an injury-free group; thus, they averaged values from both limbs in their data analyses. Both the magnitude and rate of impact loading were found to be significantly greater in their injured group vs. the injury-free group. Levinger and colleagues (2007) looked at rearfoot motion and ground reaction forces during walking in subjects with PFP and found significant differences between PFP and control groups on medial GRF magnitude and peak vertical GRF. Unfortunately, no symmetry indices were calculated. Powers et al. (1999) also looked at the influence of PFP on lower limb loading during walking and found that average peak vertical GRF of the PFP group were significantly lower than controls. Since the purpose of this study was to look at symmetry between limbs in PFP and control subjects, results of mean vertical GRF values were not reported between groups. These data could be reviewed in the future.

Our running data indicated that there was a significant group difference for maximum vertical GRF for the CHK group. The control group presented with a -3% SI, indicating greater maximum vertical GRF generated through the left limb than the right. However, this may not be clinically significant as SI values of ± 4% to 10% have been documented as "normal" limits of asymmetry for able bodied participants when assessing GRF data. Robinson and colleagues (1987) were the first to define normal gait asymmetries utilizing the symmetry index. They felt that SI values of zero to ± 10% were the upper and lower limits of normal for GRF data. Herzog et al. (1989), in a widely referenced article on asymmetries in GRF during normal gait, found a SI of 3.9% for vertical GRF. Likewise, VanZant et al. (2001), in a study of symmetry of plantar pressures and vertical forces found a similar 3.9% SI between the left and right rearfoot for maximum vertical GRF in healthy subjects. Finally, Zifchock et al. (2006) established a 10 point difference in SI as clinically significant in their kinetic study of runners with and without tibial stress fractures. Thus, the -3% SI found in our control subjects appears to be within the normal limits of symmetry for able bodied subjects as referenced in the literature. Since we did not find significant symmetry differences in our GRF data, there is no evidence to support that component of hypothesis 4.

Kinetics-COP measurement: The final biomechanical variable that we assessed was a variable we termed the “Total X-Factor”. This was a measure of mediolateral instability or stability of the COP/gait line as measured by the Pedar system. Analysis of the COP path or gait line during the stance phase of gait has been used in a variety of research studies. COP has commonly been used as a measure of temporal loading of the foot during walking. The effects of various conditions and/or treatment interventions have been studied utilizing the time that the COP falls within specific regions of the foot (Cornwall & McPoil, 2003). Another use of the COP measurement during gait has been in the investigation of foot function. Nigg et al. (2003) has proposed that the COP path is a direct result of foot pronation and supination. Based on this, Katoh and associates proposed using COP as a means of evaluating the efficacy of foot orthoses. Nigg and his group (2003) have also looked at the effects of shoe inserts on the COP during running.

To date, no one has looked at the COP/gait line variability as an indicator of lower extremity stability/instability as measured by foot function in patient populations. Hruska (1998) and the PRI have proposed that pelvifemoral dysfunction indicative of an anterior tilted, forward rotated pelvis leads to associated femoral internal rotation, medial displacement of the femoral range of rotation, genu valgus, genu recurvatum, subtalar eversion, and forefoot and/or rearfoot pronation. Clinically, patients present with left > right lower extremity instability in single leg stance as observed while performing such tasks as a lateral step down. Much of the instability pattern is subjective based on clinical observations and patient reports; objectively, we can measure step height (inches) as a means of documenting asymmetry, however, it may not be an accurate indicator of instability based on the patient's ability to compensate in other ways. Since the COP path is a direct result of foot pronation and supination during gait, the X-Factor variable could be a good indicator of lower extremity instability as indicated by the mediolateral variability of the gait line as measured along the x axis of the Pedar insole (COPx).

Based on our univariate data analysis, the SI for the "X Factor" or COPx approached significance ($p = 0.055$) with the PFP group exhibiting greater variability of the gait line on the left side as compared to the CHK group during the walking condition. In addition, while not statistically significant, the COPx also showed a trend toward more mediolateral shift of the COP/gait line of the left foot of the PFP group during the running condition. Since the Pedar system is relatively easy to use, this may be a feasible option for clinicians with the system to objectively measure foot/lower extremity instability and for patients to visualize that instability.

In our final aim, we attempted to determine if there was a set of biomechanical variables that clearly discriminated between subjects with and without PFP that could potentially compliment their individual clinical picture. In the final hypothesis, we proposed that there would be a distinct group of biomechanical variables that would help differentiate this patient population.

A logistic regression analysis of clinical and biomechanical variables revealed that during the walking condition, the TI was clearly a predictor of group membership. Age and Total COPx approached significance as predictors of group membership in the final walk model. During the running condition, age and TI were both predictors of group membership. Again, Total COPx shows a trend toward significance but did not reach statistical significance. These results seem to reinforce some of the discussions by Goble et al. (2003) and Schache et al. (2000). While the PFP group clearly exhibits more flexibility deficits and tightness patterns, these deficits do not appear to affect biomechanical outcomes until the speed of the activity increases, static flexibility is maximized and ROM and complex dynamic neuromotor patterns are stressed. Further research should be performed to look at biomechanical symmetry patterns during faster running trials.

Significance and Limitations

Significance: We were able to recruit and measure a very large sample of subjects. There was only one other PFP study that has looked at a group nearly as large as this sample. Brown et al. (2007), in unpublished data, looked at 35 subjects with low back pain or knee pain and 35 age-matched controls. They, however, looked at both men and women together. Our study appears to be one of the largest all female PFP groups measured. Each group was carefully selected based on clear inclusion and exclusion criteria. Our groups were similar in weight, height, BMI, and mileage run/hours of activity performed per week. There was no difference in leg dominance distribution between the groups with approximately 70% of both groups being right foot dominant. This is consistent within the laterality literature (Gabbard & Iteya, 1996; Porac & Coren, 1981; Sadeghi, Allard, Prince, & Labelle, 2000; Singh, 1970). However, the percentages of right lower-limb dominance reported here are slightly lower than the normal range of 75-82% reported by Gabbard, et al. (1996). We also found no significant pattern of leg dominance versus painful side in the PFP group. While the anterior cruciate ligament (ACL) literature has looked at leg dominance versus injury side in some of their etiological work (Negrete, Schick, & Cooper, 2007), this is not a common report in the PFP literature. To date, only two local studies make specific mention of this finding in their results. Neither group found significance in their analysis of dominant versus nondominant leg of injury (Cichanowski, Schmitt, Johnson, & Niemuth, 2007; Niemuth, Johnson, Myers, & Thieman, 2005).

This cohort of subjects with PFP was significantly older than the control group. This finding was contrary to Niemuth et al. (2005) who reported younger injured female runners in a similar study. Our results are not surprising, as it is a common clinical observation that as patients age their risk of PFP and other knee injuries increases. Logistic regression results of this group indicated that age was a predictor of PFP group membership during the running condition and approached significance during the walking condition.

Based on our results it was clear that our clinical scores were good predictors of PFP while the biomechanical variables, as a group, were not. The Modified Thomas/Extension Drop test, the Modified Ober/Adduction Drop test and the associated Tightness Index and Pelvic Symmetry Index were solid predictors of group membership in this sample of PFP patients. However, while not statistically significant in this study, the COPx variable may have predictive value as an evaluation tool. With the proper equipment and software capabilities, this measure would be easy to incorporate into a functional evaluation of a patient or runner. The objective, visual outputs (Figures 10 and 11) would be valuable for patient education. Further study of COP/gait line instability in a larger sample size of subjects is warranted.

Limitations. In light of the findings reported in the current study, there were several limitations that should be noted. Our clinical results, while significantly supportive of the proposed hypotheses, may only generalize to the population tested as all subjects specifically volunteered based on selective recruitment techniques. In addition, the tester was not blinded to the group membership. Lack of blinding may have produced unintentional bias during the physical examination portion of the data collection.

We cannot exclude noise of the data from skin movement artifacts associated with the pelvis and hip markers in the insignificance of our kinematic results. In addition, while we consistently calibrated our insoles, Pedar data may have been affected by sensor breakdown in older insole pairs. Finally, 30 second walk/run trials may have been too short to develop consistent, normal patterns of walking and running for each subject during the pelvis/hip kinematic studies.

One other concern about gait symmetry or asymmetry and laterality research may be the way our observations are quantified. While the symmetry index (SI) continues to be the most commonly reported equation used with kinematic and kinetic inter-limb data comparisons, there may be limitations to its use as differences are reported against their average values. For example, if a large asymmetry is present, the average value does not adequately reflect what is happening at each limb. Also, parameters that have large values but relatively small inter-limb differences will tend to lower the index and perhaps incorrectly reflect symmetry (Sadeghi et al., 2000). In spite of these limitations, the SI continues to be used while newer multivariate analyses are tested.

Conclusions

Despite a substantial amount of research on PFP during the past 25 years, it still remains the number one running injury (Taunton et al., 2002), suggesting that minimal progress has been made in prevention. Prevention is difficult if risk factors remain unknown. Witvrouw and colleagues (2000) stated that the determination of both the *biomechanical and clinical* intrinsic and extrinsic risk factors of anterior knee pain in an athletic or active population is the first step in injury prevention and treatment of PFP. In this study, a significant association was found between an increased number of positive signs on the Modified Thomas and Ober tests and PFP. These assessment tests are quick and easy to perform in a clinical setting.

Our choice of biomechanical variables, with the exception of the COPx, did not add significantly to the discrimination between groups. However, we believe that the use of the COP/gait line variability may be helpful in predicting future problems with pelvifemoral instability and could be utilized prospectively with certain subject populations in preventative assessments and screenings of athletic and active populations. We should continue to look at other biomechanical variables in future studies of PFP and pelvifemoral instability.

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Appendix A

University of Minnesota – Gait and Posture Laboratory
Patellofemoral Pain Study
Subject Information Form

Subject Number:________________________ Date:______________________________

Age:____________________ Birthdate:___________________________

Height:___________ Weight:_________ Body Mass Index:_____________________

Eligible: Yes No Agrees to Consent: Yes No

Leg Dominance: R L Handedness: R L

Shoe Size/Insole:__________

Painful Knee: R L B None Involved Limb based on PPE R LN/A

Duration of Symptoms (Knee Pain) in weeks:_________

History:____

How much (in hours per week) time do you spend practicing your sport? __________________

Mileage run per week?____________________

PPE Assessment Tests:

Thomas: L (pos. or neg.) R (pos. or neg.) Ober: L (pos. or neg.) R (pos. or neg.)

Trunk rotation: L (pos. or neg.) R (pos. or neg.)

Step Test (# performed in 30 sec.) L_____________ R_____________

Hip Rotation: L (passive IR/ER):___________ R (passive IR/ER):___________

L (active IR/ER):_____________ R (active IR/ER): ____________

SLR test (R/L) : __________ Adductor Lift Test: L___________ R_____________

Subject Accepted for study (investigator's initials):________ Date:__________

Version: 10/12/05 IRB Code:0507M71486

Appendix B

KUJALA KNEE SURVEY

Study ID: ____________________ Date: ____________________

Age: ____________ Knee: L R B

Duration of symptoms: __________ years___________months

For each question, circle the latest choice (letter) which corresponds to your knee symptoms.

1. Limp
   a. None (5)
   b. Slight or periodical (3)
   c. Constant (0)

2. Support
   a. Full support without pain (5)
   b. Painful (3)
   c. Weight bearing impossible (0)

3. Walking
   a. Unlimited (5)
   b. More than 2 km (3)
   c. 1-2 km (2)
   d. Unable (0)

4. Stairs
   a. No difficulty (10)
   b. Slight pain when descending (8)
   c. Pain both when descending and ascending (5)
   d. Unable (0)

5. Squatting
   a. No difficulty (5)
   b. Repeated squatting painful (4)
   c. Painful each time (3)
   d. Possible with partial weight bearing (2)
   e. Unable (0)

6. Running
   a. No difficulty (10)
   b. Pain after more than 2 km (8)
   c. Slight pain from start (6)
   d. Severe pain (3)
   e. Unable (0)

7. Jumping
   a. No difficulty (10)
   b. Slight difficulty (7)
   c. Constant pain (2)
   d. Unable (0)

8. Prolonged sitting with the knees flexed
   a. No difficulty (10)
   b. Pain after exercise (8)
   c. Constant pain (6)
   d. Pain forces you to extend knees temporarily
   e. Unable (0)

9. Pain
   a. None (10)
   b. Slight and occasional (8)
   c. Interferes with sleep (6)
   d. Occasionally severe (3)
   e. Constant and severe (0)

10. Swelling
    a. None (10)
    b. After severe exertion (8)
    c. After daily activities (6)
    d. Every evening (4)
    e. Constant (0)

11. Abnormal painful kneecap (patellar) movements (subluxations)
    a. None (10)
    b. Occasionally in sports activities (6)
    c. Occasionally in daily activities (4)
    d. At least one documented dislocation (2)
    e. More than two dislocations (0)

12. Atrophy of thigh
    a. None (5)
    b. Slight (3)
    c. Severe (0)

13. Flexion deficiency
    a. None (5)
    b. Slight (3)
    c. Severe (0)

Appendix C

Calculations of Pelvis and Hip Axes Systems

Three dimensional pelvis and hip joint angles were calculated utilizing a customized Matlab program suggested by the University of Calgary, and based on the International Society of Biomechanics (ISB) recommendations on definitions of joint coordinate systems (JCS) for the hip and pelvis (Wu et al., 2002). Passive reflective markers were attached to the following boney landmarks based on ISB recommendations:

- bilateral anterior superior iliac spines (ASIS)
- bilateral posterior superior iliac spines (PSIS)
- a sacral point midway between the two PSIS
- bilateral greater trochanters
- bilateral medial and lateral femoral epicondyles
- bilateral tibial tubercles
- bilateral medial and lateral tibial condyles
- the tips of the medial and lateral malleoli

These markers were used to define the embedded coordinate systems of the pelvis, hip, and thigh segments. The first step in the calculation was be get the X Y Z coordinates of each point as a vector. For example: R_PSIS=[PSISx-element y-element z-element]. This was completed for all defined points of the pelvis and hip segments:

```
L_ASIS = [L_ASIS_x L_ASIS_y L_ASIS_z];
R_ASIS = [R_ASIS_x R_ASIS_y R_ASIS_z];
R_PSIS = [R_PSIS_x R_PSIS_y R_PSIS_z];
L_PSIS = [L_PSIS_x L_PSIS_y L_PSIS_z];
R_LAT_EPI = [R_LAT_EPI_x R_LAT_EPI_y R_LAT_EPI_z];
L_LAT_EPI = [L_LAT_EPI_x L_LAT_EPI_y L_LAT_EPI_z];
R_MED_EPI = [R_MED_EPI_x R_MED_EPI_y R_MED_EPI_z];
L_MED_EPI = [L_MED_EPI_x L_MED_EPI_y L_MED_EPI_z];
R_G8_TROCH = [R_G8_TROCH_x R_G8_TROCH_y R_G8_TROCH_z];
L_G8_TROCH = [L_G8_TROCH_x L_G8_TROCH_y L_G8_TROCH_z];
```

The joint angles were calculated as ordered rotations between anatomically aligned reference frames associated with the adjacent body segments (Schutte et al., 2000). The *traditionally* defined global coordinate system follows the right hand rule and is defined as the positive x-direction oriented in the direction of forward walking or running progression, the positive y-direction oriented upward and the positive z-direction oriented laterally (Schutte et al., 2000). The standard 3-D angular rotations of the pelvis and hips are computed using a technique equivalent to the geometrical conventions described by Grood and Suntay (1983). ISB recommended pelvic motion is typically measured as rotation of the pelvic segment with respect to the global coordinate system and is isolated as one movement, pelvic tilt, without differentiation of right and left hemi-pelvic motion. Hip motion is then measured as rotation of the thigh segment with respect to that pelvic segment (Schache, Blanch, & Murphy, 2000; Wu et al., 2002). Due to frequent movement and set-up/take down of the motion capture system, we were unable to define a global coordinate system. In addition, in order to measure kinematic asymmetry of the pelvis, a modified pelvic coordinate system was developed to differentiate right versus left hemi-pelvic motion. The pelvic coordinate system was computed utilizing a modified technique which continued to follow the right-hand rule; however, the individual right and left pelvic coordinate systems were computed utilizing individualized resting standing frames recorded preliminary to each subject's movement trials. Subjects stood in their normal resting state and no adjustments to pelvic position were made. Hip motion was measured as rotation of the thigh segment with respect to the standing hemi pelvic segment on both the right and left sides. The right and left pelvic axes systems were built individually. To build the right pelvic axis:

```
Pelvis_global_right_Z(:,:) = (R_ASIS(:,:) - L_ASIS(:,:))/norm(R_ASIS(:,:) - L_ASIS(:,:)); [this will point laterally towards right]
 inter_axis (:,:)= (R_ASIS(:,:)-(R_PSIS(:,:) +L_PSIS(:,:))/2)/norm(R_ASIS(:,:)-(R_PSIS(:,:)
+L_PSIS(:,:)); [this will point anteriorly]
 Pelvis_global_right_Y(:,:) =
(cross(Pelvis_global_right_Z(:,:),inter_axis(:,:)))/norm(cross(Pelvis_global_right_Z(:,:),inter_axis(:,:)));
[this points upward]
 Pelvis_global_right_X (:,:)= (cross(Pelvis_global_right_Y(:,:),
Pelvis_global_right_Z(:,:))/norm(cross(Pelvis_global_right_Y(:,:), Pelvis_global_right_Z(:,:)); [points forward]
   Pelvis_global_right_Org (:,:)= (R_ASIS(:,:) + L_ASIS(:,:))/2;

   Pelvis_global_right = zeros(4,4);
   Pelvis_global_right(4,4) = 1;
   for i=1:3
      Pelvis_global_right(i,4) = Pelvis_global_right_Org(i,1);
   end
          for i=1:3
          Pelvis_global_right(1,i)=Pelvis_global_right_X (1,i);
```

```
      Pelvis_global_right(2,i)=Pelvis_global_right_Y (1,i);
      Pelvis_global_right(3,i)=Pelvis_global_right_Z (1,i);
```

<u>To build the left pelvic axes</u>:

Pelvis_global_left_Z(:,:) = (L_ASIS(:,:) - R_ASIS(:,:))/norm(L_ASIS(:,:) - R_ASIS(:,:)); ***[pointing laterally to the left]***;
Pelvis_global_left_Y(:,:) = (cross(inter_axis(:,:), Pelvis_global_left_Z(:,:)))/norm(cross(inter_axis(:,:), Pelvis_global_left_Z(:,:))); ***[this points upward]***
Pelvis_global_left_X(:,:) = (cross(Pelvis_global_left_Y(:,:), Pelvis_global_left_Z(:,:))/norm(cross(Pelvis_global_left_Y(:,:), Pelvis_global_left_Z(:,:)); ***[points backwards]***
Pelvis_global_left_Org (:,:)= Pelvis_global_right_Org (:,:);

```
 Pelvis_global_left = zeros(4,4);
  Pelvis_global_left(4,4) = 1;
  for i=1:3
      Pelvis_global_left(i,4) = Pelvis_global_left_Org(i,1);
  end
         for i=1:3
         Pelvis_global_left(1,i)=Pelvis_global_left_X (1,i);
      Pelvis_global_left(2,i)=Pelvis_global_left_Y (1,i);
      Pelvis_global_left(3,i)=Pelvis_global_left_Z (1,i);
```

The femoral anatomical coordinate systems were calculated in the following manner.

<u>For the right femur</u>:

Femur_anat_Right_Y(:,:) = (R_G8_TROCH(:,:))-(R_LAT_EPI(:,:)+R_MED_EPI(:,:)/2)/norm(R_G8_TROCH(:,:))-(R_LAT_EPI(:,:)+R_MED_EPI(:,:)/2); **[points upward]**
interaxis_R(:,:) = (R_LAT_EPI(:,:) - R_MED_EPI(:,:))/norm(R_LAT_EPI(:,:) - R_MED_EPI(:,:)); **[points laterally]**
Femur_anat_Right_X(:,:) = (cross(Femur_anat_Right_Y(:,:),interaxis_R(:,:))/norm(cross(Femur_anat_Right_Y(:,:),interaxis_R(:,:)); **[points forwards]**
Femur_anat_Right_Z(:,:) = (cross(Femur_anat_Right_X(:,:),Femur_anat_Right_Y(:,:))/norm(cross(Femur_anat_Right_X(:,:),Femur_anat_Right_Y(:,:)); **[points laterally right]**

```
  Femur_anat_Right_Org(:,:) = (R_LAT_EPI(:,:)+R_MED_EPI(:,:)/2);
   Femur_anat_Right(:,:) = zeros(4,4);
         Femur_anat_Right(4,4) = 1;

  for i=1:3
      Femur_anat_Right(i,4) = Femur_anat_Right_Org(i,1);
  end

  for i=1:3
         Femur_anat_Right(1,i)=Femur_anat_Right_X (1,i);
      Femur_anat_Right(2,i)=Femur_anat_Right_Y (1,i);
      Femur_anat_Right(3,i)=Femur_anat_Right_Z (1,i);
  End
```

<u>For the left femur</u>:

```
Femur_anat_Left_Y(:,:) = (L_G8_TROCH(:,:))-
(L_LAT_EPI(:,:)+L_MED_EPI(:,:)/2)/norm(L_G8_TROCH(:,:))-(L_LAT_EPI(:,:)+L_MED_EPI(:,:)/2);
[points upward]
   interaxis_L(:,:) = (L_LAT_EPI(:,:) - L_MED_EPI(:,:))/norm(L_LAT_EPI(:,:) - L_MED_EPI(:,:));
[points laterally]
   Femur_anat_Left_X(:,:) =
(cross(Femur_anat_Left_Y(:,:),interaxis_L(:,:))/norm(cross(Femur_anat_Left_Y(:,:),interaxis_L(:,:));
[points backwards]
   Femur_anat_Left_Z(:,:) =
(cross(Femur_anat_Left_X(:,:),Femur_anat_Left_Y(:,:))/norm(cross(Femur_anat_Left_X(:,:),Femur_anat
_Left_Y(:,:));  [points laterally left]
   Femur_anat_Left_Org(:,:) = (L_LAT_EPI(:,:)+L_MED_EPI(:,:)/2);

   Femur_anat_Left(:,:) = zeros(4,4);
   Femur_anat_Left(4,4) = 1;

   for i=1:3
      Femur_anat_Left(i,4) = Femur_anat_Left_Org(i,1);
   end

   for i=1:3
           Femur_anat_Left(1,i)=Femur_anat_Left_X (1,i);
      Femur_anat_Left(2,i)=Femur_anat_Left_Y (1,i);
      Femur_anat_Left(3,i)=Femur_anat_Left_Z (1,i);
   end
```

The next part of the calculation involved reading in the walk and run motion files and making a coordinate system at each frame which then found that frame's relation to stand throughout the motion.

<u>Right axis system</u>:

```
Pelvis_walk_right_Z(:,:,m) = (R_ASIS(:,:,m) - L_ASIS(:,:,m))/norm(R_ASIS(:,:,m) - L_ASIS(:,:,m));
[this will point laterally towards right]
       inter_axis (:,:,m)= (R_ASIS(:,:,m)-(R_PSIS(:,:,m) +L_PSIS(:,:,m))/2)/norm(R_ASIS(:,:,m)-
(R_PSIS(:,:,m) +L_PSIS(:,:,m));  [this will point anteriorly]
       Pelvis_walk_right_Y(:,:,m) =
(cross(Pelvis_walk_right_Z(:,:,m),inter_axis(:,:,m)))/norm(cross(Pelvis_walk_right_Z(:,:,m),inter_axis(:,:
,m)));  [this points upward]
       Pelvis_walk_right_X (:,:,m)= (cross(Pelvis_walk_right_Y(:,:,m),
Pelvis_walk_right_Z(:,:,m))/norm(cross(Pelvis_walk_right_Y(:,:,m), Pelvis_walk_right_Z(:,:,m));
[points forward]
       Pelvis_walk_right_Org (:,:,m)= (R_ASIS(:,:,m) + L_ASIS(:,:,m))/2;

       Pelvis_walk_right = zeros(4,4);
       Pelvis_walk_right(4,4,m) = 1;
       for i=1:3
          Pelvis_walk_right(i,4,m) = Pelvis_walk_right_Org(i,1);
       end
       for i=1:3
          Pelvis_walk_right(1,i,m)=Pelvis_walk_right_X (1,i);
```

```
    Pelvis_walk_right(2,i,m)=Pelvis_walk_right_Y (1,i);
    Pelvis_walk_right(3,i,m)=Pelvis_walk_right_Z (1,i);
End
```

Left axis system:

Pelvis_walk_left_Z(:,:,m) = (L_ASIS(:,:,m) - R_ASIS(:,:,m))/norm(L_ASIS(:,:,m) - R_ASIS(:,:,m)); **[pointing laterally to the left]**

Pelvis_walk_left_Y(:,:,m) = (cross(inter_axis(:,:,m), Pelvis_walk_left_Z(:,:,m)))/norm(cross(inter_axis(:,:,m), Pelvis_walk_left_Z(:,:,m))); **[this points upward]**

Pelvis_walk_left_X(:,:,m) = (cross(Pelvis_walk_left_Y(:,:,m), Pelvis_walk_left_Z(:,:,m))/norm(cross(Pelvis_walk_left_Y(:,:,m), Pelvis_walk_left_Z(:,:,m)); **[points backwards]**

Pelvis_walk_left_Org (:,:,m)= Pelvis_walk_right_Org (:,:,m);

```
Pelvis_walk_left = zeros(4,4);
Pelvis_walk_left(4,4,m) = 1;
for i=1:3
    Pelvis_walk_left(i,4,m) = Pelvis_walk_left_Org(i,1);
end
for i=1:3
    Pelvis_walk_left(1,i,m)=Pelvis_walk_left_X (1,i);
    Pelvis_walk_left(2,i,m)=Pelvis_walk_left_Y (1,i);
    Pelvis_walk_left(3,i,m)=Pelvis_walk_left_Z (1,i);
end
```

Right femur walk file:

Femur_walk_Right_Y(:,:,m) = (R_G8_TROCH(:,:,m))-(R_LAT_EPI(:,:,m)+R_MED_EPI(:,:,m)/2)/norm(R_G8_TROCH(:,:,m))-(R_LAT_EPI(:,:,m)+R_MED_EPI(:,:,m)/2); **[points upward]**

interaxis_R(:,:,m) = (R_LAT_EPI(:,:,m) - R_MED_EPI(:,:,m))/norm(R_LAT_EPI(:,:,m) - R_MED_EPI(:,:,m)); **[points laterally]**

Femur_walk_Right_X(:,:,m) = (cross(Femur_walk_Right_Y(:,:,m),interaxis_R(:,:,m))/norm(cross(Femur_walk_Right_Y(:,:,m),interaxis_R(:,:,m)); **[points forward]**

Femur_walk_Right_Z(:,:,m) = (cross(Femur_walk_Right_X(:,:,m),Femur_walk_Right_Y(:,:,m))/norm(cross(Femur_walk_Right_X(:,:,m),Femur_walk_Right_Y(:,:,m)); **[points laterally to right]**

Femur_walk_Right_Org(:,:,m) = (R_LAT_EPI(:,:,m)+R_MED_EPI(:,:,m)/2);

```
Femur_walk_Right(:,:,m) = zeros(4,4);
Femur_walk_Right(4,4,m) = 1;

for i=1:3
    Femur_walk_Right(i,4,m) = Femur_walk_Right_Org(i,1);
end

for i=1:3
    Femur_walk_Right(1,i,m)=Femur_walk_Right_X (1,i);
    Femur_walk_Right(2,i,m)=Femur_walk_Right_Y (1,i);
    Femur_walk_Right(3,i,m)=Femur_walk_Right_Z (1,i);
End
```

Left femur walk file:

```
Femur_walk_Left_Y(:,:,m) = (L_G8_TROCH(:,:,m))-
(L_LAT_EPI(:,:,m)+L_MED_EPI(:,:,m)/2)/norm(L_G8_TROCH(:,:,m))-
(L_LAT_EPI(:,:,m)+L_MED_EPI(:,:,m)/2);  [points upward]
interaxis_L(:,:,m) = (L_LAT_EPI(:,:,m) - L_MED_EPI(:,:,m))/norm(L_LAT_EPI(:,:,m) -
L_MED_EPI(:,:,m));  [points laterally]
Femur_walk_Left_X(:,:,m) =
(cross(Femur_walk_Left_Y(:,:,m),interaxis_L(:,:,m))/norm(cross(Femur_walk_Left_Y(:,:,m),interaxis_L
(:,:,m));  [points backwards]
Femur_walk_Left_Z(:,:,m) =
(cross(Femur_walk_Left_X(:,:,m),Femur_walk_Left_Y(:,:,m))/norm(cross(Femur_walk_Left_X(:,:,m),F
emur_walk_Left_Y(:,:,m));  [points laterally to left]
Femur_walk_Left_Org(:,:,m) = (L_LAT_EPI(:,:,m)+L_MED_EPI(:,:,m)/2);

Femur_walk_Left(:,:,m) = zeros(4,4);
Femur_walk_Left(4,4,m) = 1;

for i=1:3
   Femur_walk_Left(i,4,m) = Femur_walk_Left_Org(i,1);
end

for i=1:3
   Femur_walk_Left(1,i,m)=Femur_walk_Left_X (1,i);
   Femur_walk_Left(2,i,m)=Femur_walk_Left_Y (1,i);
   Femur_walk_Left(3,i,m)=Femur_walk_Left_Z (1,i);
end
```

Next we found the transformation of the resting femur position with the resting pelvic position.

```
Femur_rest_R(:,:) = (Pelvis_anat_right(:,:)^-1)*Femur_anat_Right(:,:);
 Femur_rest_L(:,:) = (Pelvis_anat_left(:,:)^-1)*Femur_anat_Left(:,:);
```

To find the transformation of each walk (or run) frame with the anatomical transformation the following was run:

```
for m=1:length(A(:,1))
   Pelvic_trans_R(:,:,m) = (Pelvis_anat_right(:,:)^-1)*Pelvis_walk_right(:,:,m);
   Pelvic_trans_L(:,:,m) = (Pelvis_anat_left(:,:)^-1)*Pelvis_walk_left(:,:,m);
   Femur_trans_R(:,:,m) = (Pelvis_walk_Right(:,:,m)^-1)*Femur_walk_Right(:,:,m);
   Femur_trans_L(:,:,m) = (Pelvis_walk_Left(:,:,m)^-1)*Femur_walk_Left(:,:,m);
end
```

Euler angles using a ZXY sequence for the resting stand position between femur and pelvis were then extracted.

```
Femur_rest_R_alpha(:,:) = atan2(-Femur_rest_R(1,2),Femur_rest_R(2,2));
Femur_rest_R_beta(:,:) = atan2(Femur_rest_R(3,2),sqrt(Femur_rest_R(1,2)^2+Femur_rest_R(2,2)^2));
Femur_rest_R_gamma(:,:) = atan2(-Femur_rest_R(1,3),Femur_rest_R(3,3));
```

```
Femur_rest_L_alpha(:,:) = atan2(-Femur_rest_L(1,2),Femur_rest_L(2,2));
Femur_rest_L_beta(:,:) = atan2(Femur_rest_L(3,2),sqrt(Femur_rest_L(1,2)^2+Femur_rest_L(2,2)^2));
Femur_rest_L_gamma(:,:) = atan2(-Femur_rest_L(1,3),Femur_rest_L(3,3));
```

Euler angles using a ZXY sequence for the motion file between the femur and pelvis were also extracted.

```
Femur_trans_R_alpha(:,:,m) = atan2(-Femur_trans_R(1,2,m),Femur_trans_R(2,2,m));
Femur_trans_R_beta(:,:,m) =
atan2(Femur_trans_R(3,2,m),sqrt(Femur_trans_R(1,2,m)^2+Femur_trans_R(2,2,m)^2));
 Femur_trans_R_gamma(:,:,m) = atan2(-Femur_trans_R(1,3,m),Femur_trans_R(3,3,m));

 Femur_trans_L_alpha(:,:,m) = atan2(-Femur_trans_L(1,2,m),Femur_trans_L(2,2,m));
 Femur_trans_L_beta(:,:,m) =
atan2(Femur_trans_L(3,2,m),sqrt(Femur_trans_L(1,2,m)^2+Femur_trans_L(2,2,m)^2));
 Femur_trans_L_gamma(:,:,m) = atan2(-Femur_trans_L(1,3,m),Femur_trans_L(3,3,m));
```

The variables were then renamed to isolate the variables used in this data analysis.

```
walk_FlexExt_Femur_R = (Femur_trans_angles_R(:,1));
walk_FlexExt_Femur_L = (-1)*(Femur_trans_angles_L(:,1));
walk_FlexExt_Pelvis_R = (Pelvis_trans_angles_R(:,1));
walk_FlexExt_Pelvis_L = (-1)*(Femur_trans_angles_L(:,1));
```

We then isolated the means, maximum and minimum values of the desired data during the 30 second walk or run trials.

```
walk_mean_FlexExt_Femur_R = mean(walk_FlexExt_Femur_R(i,1))
walk_mean_FlexExt_Femur_L = mean(walk_FlexExt_Femur_L(i,1))
walk_mean_FlexExt_Pelvis_R = mean(walk_FlexExt_Pelvis_R(i,1))
walk_mean_FlexExt_Pelvis_L = mean(walk_FlexExt_Pelvis_L(i,1))

 NoStepsPerTrial = 30
     a = LOF_WalkData/NoStepsPerTrial;
        for b = 1:a;
             c = b*20;
             d = c-19;
          Walk_Flex_Femur_R_Max(b,1) = max(walk_FlexExt_Femur_R(d:c,1))
          Walk_Flex_Femur_L_Max(b,1) = max(walk_FlexExt_Femur_L(d:c,1))
          Walk_Flex_Pelvis_R_Max(b,1) = max(walk_FlexExt_Pelvis_R(d:c,1))
          Walk_Flex_Pelvis_L_Max(b,1) = max(walk_FlexExt_Pelvis_L(d:c,1))
          Walk_Ext_Femur_R_Max(b,1) = min(walk_FlexExt_Femur_R(d:c,1))
          Walk_Ext_Femur_L_Max(b,1) = min(walk_FlexExt_Femur_L(d:c,1))
          Walk_Ext_Pelvis_R_Max(b,1) = min(walk_FlexExt_Pelvis_R(d:c,1))
          Walk_Ext_Pelvis_L_Max(b,1) = min(walk_FlexExt_Pelvis_L(d:c,1))
      end
```

Appendix D

Gait and Posture Laboratory

UNIVERSITY OF MINNESOTA

CONSENT FORM SUBJECT
Patellofemoral Pain Study

You are invited to participate in a research study assessing the biomechanical variables associated with anterior knee pain in female athletes. You were selected as a possible participant because you have persistent anterior knee pain. We ask that you read this form and ask any questions you may have before agreeing to be in the study.

This study is being conducted by Karen Swanson, MA, ATC, ATR, PT and Juergen Konczak, PhD, School of Kinesiology, University of Minnesota.

Study Purpose

The purpose of the study is to assess pelvis and lower extremity alignment in female anterior knee pain patients and healthy control subjects through gait analysis and force pressure measures. The study will attempt to establish a biomechanical profile of female Patellofemoral pain patients.

Study Procedures

If you agree to participate in this study, we would ask you to do the following:

1. Report to the Gait and Posture Laboratory for 1 testing session. At this visit, you will be asked to answer demographic questions such as age, height, weight, and athletic participation level. You will also undergo a clinical assessment of pelvis and lower extremity alignment, and other postural measurements. Following the clinical assessment, a functional step test and gait analysis will be performed. You will be asked to place a pair of insoles into your shoes to measure force pressures. You will have numerous reflective markers placed on your pelvis, hips, thighs, lower legs and feet in order to collect data during the gait analysis. During the analysis, you will be asked to perform numerous activities such as a single leg step test on a block, single and double leg stance, walking on a treadmill and running on a treadmill. Finally, you will be asked to fill out 2 questionnaires in which you will subjectively rate your knee(s) and a visual pain scale before and after the step test. Total time for these tests should not exceed 75 minutes.

Risks of Study Participation

The study has the following risks: There is minimal risk associated with participation in this study. There is always the routine risk that you might injure yourself while walking or running on the treadmill or while performing the functional step test. Note that the potential risks are no greater than those that would be encountered in standard training or exercise sessions or during daily function. There is also a minor risk of skin irritation from the adhesive backing used to secure the placement markers for the gait analysis.

Benefits of Study Participation

There are no anticipated benefits to participation in the study. Upon request, at the end of the study, you may receive results of your clinical assessments and gait analysis.

Alternatives to Study Participation

If you refuse to participate at any time, we will not use the results of your collected data.

Study Costs/Compensation

You will not receive payment for participation.

Research Related Injury

In the event that this research activity results in an injury, treatment will be available, including first aid, emergency treatment and follow-up care as needed. Care for such injuries will be billed in the ordinary manner to you or your insurance company. If you think that you have suffered a research related injury, let the study investigators know right away.

Confidentiality

The records of this study will be kept private. In any publications or presentations, we will not include any information that will make it possible to identify you as a subject. Research records will be kept in locked files at the laboratory. Only the researchers involved with this study will have access to the records.

Voluntary Nature of the Study

Participation in this study is voluntary. Your decision whether or not to participate in this study will not affect your current or future relations with the University, Fairview-University Medical Center, your treating clinic or the researchers. If you decide to participate, you are free to withdraw at any time without affecting those relationships.

Contacts and Questions

The researchers conducting this study are Karen Swanson, MA, ATC, PT and Juergen Konczak, PhD. You may ask any questions you have now, or if you have questions later, **you are encouraged to** contact them at 612-625-3313 (Lab number); or 612-624-4370 (Dr. Konczak's office).

If you have any questions or concerns regarding the study and would like to talk to someone other than the researcher(s), **you are encouraged to** contact the Fairview Research Helpline at telephone number 612-672-7692 or toll free at 866-508-6961. You may also contact this office in writing or in person at Fairview University Medical Center - Riverside Campus, #815 Professional Building, 2450 Riverside Avenue, Minneapolis, MN 55454.

You will be given a copy of this form to keep for your records.

Statement of Consent

I have read the above information. I have asked questions and have received answers. I consent to participate in the study.

Signature of Subject____________________________________ Date__________________

Signature of Investigator________________________________ Date__________________

IRB Code: 0507M71486 Version Date: 10/12/05

Appendix E

Gait and Posture Laboratory
UNIVERSITY OF MINNESOTA

CONSENT FORM CONTROLS
Patellofemoral Pain Study

You are invited to participate in a research study assessing the biomechanical variables associated with anterior knee pain in female athletes. You were selected as a possible participant because you are an age-matched control subject without knee pain or lower extremity symptoms. We ask that you read this form and ask any questions you may have before agreeing to be in the study.

This study is being conducted by Karen Swanson, MA, ATC, ATR, PT and Juergen Konczak, PhD, School of Kinesiology, University of Minnesota.

Study Purpose

The purpose of the study is to assess pelvis and lower extremity alignment in female anterior knee pain patients and healthy control subjects through gait analysis and force pressure measures. The study will attempt to establish a biomechanical profile of female Patellofemoral pain patients.

Study Procedures

If you agree to participate in this study, we would ask you to do the following:

1. Report to the Gait and Posture Laboratory for 1 testing session. At this visit, you will be asked to answer demographic questions such as age, height, weight, and athletic participation level. You will also undergo a clinical assessment of pelvis and lower extremity alignment, and other postural measurements. Following the clinical assessment, a functional step test and gait analysis will be performed. You will be asked to place a pair of insoles into your shoes to measure force pressures. You will have numerous reflective markers placed on your pelvis, hips, thighs, lower legs and feet in order to collect data during the gait analysis. During the analysis, you will be asked to perform numerous activities such as a single leg step test on a block, single and double leg stance, walking on a treadmill and running on a treadmill. Finally, you will be asked to fill out 2 questionnaires in which you will subjectively rate your knee(s) and a visual pain scale before and after the step test. Total time for these tests should not exceed 75 minutes.

Risks of Study Participation

The study has the following risks: First, there is minimal risk associated with participation in this study. There is always the routine risk that you might injure yourself while walking or running on the treadmill or while performing the functional step test. Note that the potential risks are no greater than those that would be encountered in standard training or exercise sessions or during daily function. There is also a minor risk of skin irritation from the adhesive backing used to secure the placement markers for the gait analysis.

Benefits of Study Participation

There are no anticipated benefits to participation in the study. Upon request, at the end of the study, you may receive results of your clinical assessments and gait analysis.

Alternatives to Study Participation

If you refuse to participate at any time, we will not use the results of your collected data.

Study Costs/Compensation

You will not receive payment for participation.

Research Related Injury

In the event that this research activity results in an injury, treatment will be available, including first aid, emergency treatment and follow-up care as needed. Care for such injuries will be billed in the ordinary manner to you or your insurance company. If you think that you have suffered a research related injury, let the study investigators know right away.

Confidentiality

The records of this study will be kept private. In any publications or presentations, we will not include any information that will make it possible to identify you as a subject. Research records will be kept in locked files at the laboratory. Only the researchers involved with this study will have access to the records.

Voluntary Nature of the Study

Participation in this study is voluntary. Your decision whether or not to participate in this study will not affect your current or future relations with the University of Minnesota or the researchers. If you decide to participate, you are free to withdraw at any time without affecting those relationships.

Contacts and Questions

The researchers conducting this study are Karen Swanson, MA, ATC, PT and Juergen Konczak, PhD. You may ask any questions you have now, or if you have questions later, you are encouraged to contact them at 612-625-3313 (Lab number); or 612-624-4370 (Dr. Konczak's office).

If you have any questions or concerns regarding the study and would like to talk to someone other than the researcher(s), you are encouraged to contact the Fairview Research Helpline at telephone number 612-672-7692 or toll free at 866-508-6961. You may also contact this office in writing or in person at Fairview University Medical Center - Riverside Campus, #815 Professional Building, 2450 Riverside Avenue, Minneapolis, MN 55454.

You will be given a copy of this form to keep for your records.

Statement of Consent

I have read the above information. I have asked questions and have received answers. I consent to participate in the study.

Signature of Subject____ Date_________________

Signature of Investigator_______________________________________ Date__________________

IRB Code: 0507M71486 Version Date: 10/12/05

Appendix F

EXTENSION DROP TEST/MODIFIED THOMAS TEST

The patient is positioned in supine with both thighs on the table. Both hips and knees are flexed to the chest. Passively lower one leg over the edge of the table while helping the patient hold the untested knee close enough to the chest to maintain the low back against the table. Do not allow hip abduction to occur past zero degrees on the tested extremity while passively dropping the FA joint into extension.

A positive test is indicated when the tested lower extremity (usually the left) is restricted in hip extension because of the forward orientation of the tested side compared to the other. If both femurs do not approach the edge of the mat or table the patient is tested on, the innominates are rotated forward bilaterally and the psoas muscles are on slack. Placing the femur in “neutral” is actually placing the patient’s femur in external rotation. This tightens the TFL and VL and restricts hip extension.

There is also a rotary component to this issue, especially seen with limitation in hip extension on one side. Since the forward, anteriorly rotated pelvis accompanies sacral rotation to the contralateral side (right rotation on a right oblique axis or left rotation on a left oblique axis) the iliofemoral ligament will also limit extension when the femur is externally rotated by the therapist, through testing with the femur in a “neutral” position.

The femur in this case will not approach the patient support surface without femoral internal rotation and or through luxation (i.e. “click”) of anterior superior femoral head moving forward under the superior anterior condyloid labral rim of acetabulum.

Negative Right Extension Drop Test

Positive Left Extension Drop Test

Appendix G

ADDUCTION DROP TEST/MODIFIED OBER TEST

The patient lies on his or her side with the lower leg and hip flexed (90 degrees). Stand behind the patient and passively flex, abduct and extend the hip to neutral while maintaining 90 degrees of knee flexion. Passively stabilize the pelvis from falling backward and allowing femoral internal rotation to occur.

A positive test is indicated by a restriction from the anterior-inferior acetabular labral rim, transverse ligament, and piriformis muscle or impact of the posterior inferior femoral neck on posterior inferior rim of acetabulum that does not allow the femur to adduct; possibly secondary to an anteriorly rotated, forward hemipelvis. Usually seen on the left especially if left Extension Drop Test is positive in a Left AIC oriented patient.

Negative Right Adduction Drop Test

Positive Left Adduction Drop Test

Appendix H

Clinical Symmetry Indices

Two clinical “indices” were established based on results of the Modified Thomas and Modified Ober tests. Results of these tests were documented as either “positive” or “negative” (see Appendices E & F).

Tightness Index (TI)

This index was based on a more conventional approach to flexibility of the lower extremities. Traditionally, a Thomas test is used to measure hip flexor flexibility and the Ober test is used as a measure of iliotibial band (ITB) and lateral hip flexibility (Ober & Peltier, 1987).

The TI assessment scale is based on a measure of normal to maximal tightness with no concern for whether the positive result occurs on the right or left side.

The TI is based on a 0-4 scale:

0 = four negative tests; normal flexibility

1 = one positive test

2 = two positive tests

3 = three positive tests

4 = 4 positive tests; maximal tightness

Pelvic Symmetry Index (PSI)

The Postural Restoration Institute has described the Anterior Interior Chain (AIC) pattern of kinetic and kinematic chain dysfunction that relates to back and knee pain. This pattern of pelvifemoral instability is described as an anteriorly tilted, forwardly rotated ***left*** innominate with an associated left femur that has moved into a passive internally rotated position. This position would create the following problems relative to our clinical assessment tests (Hruska & Joutras, 1999):

- Hypertonic positional shortened hip flexors (positive Modified Thomas test)
- Forward pelvis on the left causing altered Tensor fascia latae/iliotibial band position (positive Modified Ober test)
- Hypertonic lengthened hamstrings (bicep femoris, semitendinosus) (SLR assessment)
- Passive internal rotation of the femur with accompanying internal rotation weakness and increased demands on the vastus lateralis muscle and increased lateral compressive forces on the patella

Based on this pattern/presentation, right and left sided results now become relevant. A positive left Modified Thomas, left Modified Ober and a reduced SLR on the left are indicative of a left anteriorly tipped, forward rotated pelvis and left internally rotated femur. Thus, the PSI was based on positive results of the two tests and on which side they occurred. The PSI was based on a scale of 0-3:

0 = four negative tests

1 = positive left Ober and/or left Thomas

2 = positive left and right Ober or Thomas

3 = four positive tests; bilateral left and right Ober and Thomas

Appendix I

Raw Data Descriptive Statistics (Walk)

ID	Variable	Mean	SD
CHK	SLR (R) deg	90.53	12.81
	SLR (L) deg	89.56	13.46
	Contact time (R) ms	547.64	82.49
	Contact time (L) ms	539.35	88.10
	Max Force (R) % BW	110.07	10.04
	Max Force (L) % BW	109.10	9.71
	Total X Factor (R) mm	14.41	4.69
	Total X Factor (L) mm	13.37	2.89
	Pelvic Flexion (R) deg	5.44	3.48
	Pelvic Flexion (L) deg	5.56	3.41
	Femur Extension (R) deg	11.03	6.04
	Femur Extension (L) deg	11.20	5.62
	SD Forefoot (R) mm	5.85	1.19
	SD Forefoot (L) mm	6.10	1.23
	SD Midfoot (R) mm	4.71	1.50
	SD Midfoot (L) mm	4.54	1.35
	SD Heel (R) mm	3.85	3.50
	SD Heel (L) mm	2.73	1.70
PFP	SLR (R) deg	89.79	16.53
	SLR (L) deg	90.53	18.57
	Contact time (R) ms	561.38	103.05
	Contact time (L) ms	540.66	96.67
	Max Force (R) % BW	108.54	13.62
	Max Force (L) % BW	107.74	14.49
	Total X Factor (R) mm	13.60	3.89
	Total X Factor (L) mm	14.51	3.36
	Pelvic Flexion (R) deg	0.59	4.98
	Pelvic Flexion (L) deg	0.63	4.51
	Femur Extension (R) deg	14.60	12.19
	Femur Extension (L) deg	11.96	9.45
	SD Forefoot (R) mm	6.07	1.66
	SD Forefoot (L) mm	6.46	1.28
	SD Midfoot (R) mm	4.58	1.83
	SD Midfoot (L) mm	4.66	1.16
	SD Heel (R) mm	2.95	2.06
	SD Heel (L) mm	3.39	2.37

Appendix J

Raw Data Descriptive Statistics (Run)

ID	Variable	Mean	SD
CHK	SLR (R) deg	90.53	12.81
	SLR (L) deg	89.56	13.46
	Contact time (R) ms	267.98	70.91
	Contact time (L) ms	236.06	42.25
	Max Force (R) % BW	203.08	23.49
	Max Force (L) % BW	209.40	26.36
	Total X Factor (R) mm	12.35	4.94
	Total X Factor (L) mm	12.41	3.39
	Pelvic Flexion (R) deg	-0.43	2.91
	Pelvic Flexion (L) deg	-0.04	2.74
	Femur Extension (R) deg	19.75	7.57
	Femur Extension (L) deg	18.43	10.90
	SD Forefoot (R) mm	4.71	1.44
	SD Forefoot (L) mm	5.57	1.55
	SD Midfoot (R) mm	4.14	1.93
	SD Midfoot (L) mm	3.89	1.46
	SD Heel (R) mm	3.50	3.04
	SD Heel (L) mm	2.95	2.00
PFP	SLR (R) deg	89.79	16.53
	SLR (L) deg	90.53	18.57
	Contact time (R) ms	267.32	43.84
	Contact time (L) ms	252.70	46.05
	Max Force (R) % BW	195.79	27.09
	Max Force (L) % BW	194.87	29.20
	Total X Factor (R) mm	11.39	4.32
	Total X Factor (L) mm	13.83	4.22
	Pelvic Flexion (R) deg	-3.26	4.48
	Pelvic Flexion (L) deg	-3.26	4.72
	Femur Extension (R) deg	23.75	8.68
	Femur Extension (L) deg	23.26	9.11
	SD Forefoot (R) mm	5.37	2.48
	SD Forefoot (L) mm	6.33	2.35
	SD Midfoot (R) mm	4.02	1.49
	SD Midfoot (L) mm	4.30	1.82
	SD Heel (R) mm	2.25	1.89
	SD Heel (L) mm	3.53	2.90