2006

Kinematic and Kinetic Differences in the Trunk, Pelvis, and Lower Extremities in Women With and Without Patellofemoral Pain When Descending Stairs

H. James Phillips
Seton Hall University

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KINEAMIC AND KINETIC DIFFERENCES OF THE TRUNK, PELVIS, AND LOWER EXTREMITIES IN WOMEN WITH AND WITHOUT PATELLOFEMORAL PAIN WHEN DESCENDING STAIRS

BY

Howard James Phillips

Dissertation Committee:

Dr. Lee Cabell, Chair
Dr. Diana Glendinning
Dr. Christopher Powers

Submitted in partial fulfillment of the Requirements for the Degree of Doctor of Philosophy in Health Sciences
Seton Hall University
2006
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Approved by the Dissertation Committee:

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Special thanks to Maryann Clark and Genevieve Zipp who recognized my potential for academic pursuits over 10 years ago...about 10 years before I realized the same potential!
Dedication

This work is dedicated to my wife Laura Ngan Phillips and to my sons, Conor, Carl, and Cameron, who spent many evenings and weekends without their Dad who was working on "just one more" project for school or work. Their love and support makes it all worthwhile.
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KINEMATIC AND KINETIC DIFFERENCES OF THE TRUNK, PELVIS, AND LOWER EXTREMITIES IN WOMEN WITH AND WITHOUT PATELLOFEMORAL PAIN WHEN DESCENDING STAIRS

Howard James Phillips
Seton Hall University
2006

Chair: Dr. Lee Cabell

It is known that altered knee mechanics in individuals with PFP during descent of stairs produces increased joint stress and pain. Some theories suggest that altered femoral alignment may be a leading cause, possibly originating from weak hip musculature and differences in pelvic girdle alignment. The purpose of the study was to determine if kinematic and kinetic differences exist in the knee, hip, and pelvis between women with and without patellofemoral pain (PFP) when descending stairs. Ten female subjects with PFP and ten age-matched female subjects without PFP were studied. Each descended a standard staircase while knee, hip, and pelvis kinematics, and ground reaction force were obtained. Joint angles and moments were calculated using inverse dynamics equations. Using Mann-Whitney U non-parametric statistical analyses, PFP subjects had increased knee valgus moments (p = .006) increased hip abduction moments (p = .001), and increased ipsilateral pelvis obliquity angles (p = .0001), compared to control subjects. Further, pelvic obliquity angles correlated with knee valgus moments (r = .44, p = .027). These findings among PFP subjects suggest that the differences seen in their knee biomechanics may originate from pelvic girdle alignment, most likely due to differences in hip strength. The association between the pelvis, hip and knee requires further investigation to better understand the underlying biomechanical causes of patellofemoral pain.
CHAPTER I
INTRODUCTION

Patellofemoral pain (PFP), described as anterior knee or retropatellar discomfort, usually brought on with activity that loads the knee extensor mechanism (Greisamer & McConnell, 1998), as shown in Figure 1, is one of the most common knee pain complaints seen in sports medicine clinics, (Devereaux & Lachmann, 1984) in orthopedic practice, (Fulkerson, 2002; Henry, 1998) family practice (Juhn, 1996) and physical therapy clinics (Steinramp, Dillingham, Markel, Hill, & Kaufman, 1993; Powers 1998; Malone, Davies & Walsh, 2002). Despite its prevalence the cause of patellofemoral pain is poorly understood (Grabiner, Koh & Draganich, 1994; Greisamer & McConnell 1998; Powers 1998; Schreyer, Labs, & Wagner, 2001). The most common explanation is increased patellofemoral joint stress (joint reaction force per unit area) (Goodfellow, Hungerford & Woods, 1976) due to reduced bony joint surface area from morphologic variability (Powers, 2000) or abnormal patellar tracking (Doucette & Goble, 1992). For many PFP patients, the patella tracks laterally in the femoral trochlea during knee extensor loading, with joint reaction forces distributed over a reduced surface area compared to pain-free individuals, as shown in Figure 2. As such, interventions aimed at relieving the malady have focused on improving alignment and tracking of
Figure 1. Free body diagram of forces acting on knee joint when extensor mechanism is loaded. $F$ = internal joint moment causing flexion of knee; $M_e$ = external muscle moment in response to internal joint moment; $R_s$ = resultant patellofemoral joint reaction force.

Figure 2. Normal patellofemoral contact, left, and reduced patellofemoral contact area due to lateral tracking and compression.

A major focus of attention in the conservative management of PFP by physical therapists has been on strengthening of the quadriceps musculature, particularly the vastus medialis oblique (VMO) to control patella tracking (Doucette & Goble, 1992; Banovetz, Banovetz, & Albright, 1998; Callaghan & Oldham, 1996; Powers, et al. 1996; Grelsamer & McConnell, 1998). A consensus has arisen that overall strength of the quadriceps is required to counteract the normal tendency of the patella to track laterally due to bony geometry. This popular approach has produced generally good results in relieving PFP symptoms (Banovetz, et al. 1996).

Despite the widespread acceptance and positive results following quadriceps strengthening, some patellofemoral pain patients continue to
experience patellofemoral pain despite seemingly normal quadriceps strength. For some of these individuals, clinicians have suggested that the ipsilateral femur assumes a more adducted and/or internally rotated position during knee extensor mechanism loading, such as during midstance of gait and weight-acceptance during descent of stairs (Powers, 2003). For these individuals the femur appears to be mal-aligned with the patella which is tethered within the extensor mechanism.

The first evidence to support this theory is magnetic resonance imaging (MRI) taken during functional loading of the knee joint. Images produced by Powers (2003) revealed a demonstrable internal rotation of the femur beneath the relatively stationary patella, resulting in lateral tilt and/or subluxation of the patella during a squatting maneuver, as shown in Figures 3 and 4. While this MRI evidence is compelling, no kinematic studies of femoral rotation among PFP subjects was found.

These observations of the femur moving beneath the relatively stationary patella have lead to a "paradigm shift" in clinicians’ and researchers’ focus, with attention being drawn to control of the femur beneath the patella, rather than control of the patella atop the femur (Powers, 2003) for this subset of PFP patients.
Figure 3. MRI showing normal alignment of patellofemoral joint prior to functional loading of joint. (Reproduced with permission from Powers, C.M. Patellofemoral Disorders: Home Study Course, APTA, ©2000 CM Powers.)

Figure 4. MRI showing lateral subluxation of patellofemoral joint during functional loading of joint due to internal rotation of femur. (Reproduced with permission from Powers, C.M., Patellofemoral Disorders: Home Study Course, APTA. © 2000 C.M. Powers.)
With respect to femoral alignment, two general theories have evolved: A) that femoral adduction and rotation is controlled by the hip musculature from above, which may have a pelvic girdle or "core" strength component; the so-called "top down" theory or B) that femoral adduction and rotation is largely determined by alignment of the tibia distally, which may be influenced by foot alignment (pronation,) known as the "bottom up" theory, (Powers, 2003), as shown in Figure 5.

Figure 5. A- Influence of pelvis resulting from or contributing to hip abductor weakness and B- influence of foot pronation causing internal rotation of tibia and femur, altering knee alignment. (Modified from Powers, CM. The influence of altered lower extremity kinematics on patellofemoral joint dysfunction: a theoretical perspective. J Orthop Exam Sports Phys Ther. 2003; 33(11):639-646, with permission of the Orthopedic and Sports Section of the American Physical Therapy Association.)
In the top-down theory, weak hip abductors allow the 'thigh to adduct with respect to the pelvis, causing a compensatory abduction of the tibia, ultimately resulting in increased knee valgus (Powers, 2003). This increased knee valgus has been linked to patellofemoral pain and osteoarthritic changes of the joint (Cahue, Dunlop, Hayes, Song, Torres, & Sharma, 2004). However, no studies have measured frontal plane motions of the knee during dynamic activities to test this theoretical assertion.

The top down approach has more recently been expanded to look further up the biomechanical chain, to include consideration of how the pelvis may influence femoral alignment since the pelvis forms the basis or "core" for hip muscle attachments, (Mascal, et al., 2003). In this theory, the considerable external hip adduction moment produced by the superincumbent body weight during gait or single-limb stance needs to be balanced by the hip abductor muscles. If the hip abductors are weak, a resulting tilt of the pelvis to either the stance or opposite side may result (Perry, 1992) and, more importantly, will allow the thigh to adduct or "collapse" (Powers, 2003) leading to increased knee valgus moment and angle. This medial displacement of the knee will cause lateral tracking of the patella since it is tethered between the relatively stationary pelvis from above, and the tibial tubercle from below (Figure 6).
Figure 6. Diagram showing external hip adduction moment and external knee valgus moment in response to ground reaction force (red arrow) causing lateral tracking of patella.
The bottom up approach, while having a frequently cited theoretical basis of increased foot pronation leading to faulty knee alignment (Tiberio, 1987), has had only sparse scientific investigation among patellofemoral pain subjects. Messner, Curl, Davis, Lowery and Pack (1991) found no correlation between peak pronation motion and velocity, or rearfoot motion, in patellofemoral pain and pain-free subjects. Powers, Chen, Reischl and Perry (2002) found no correlation between the magnitude and timing of foot pronation and tibial internal rotation among both PFP and pain-free subjects.

Finally, attention has turned to the influence of trunk alignment and patellofemoral pain. Researchers have noted reduced knee flexion moments in PFP patients when descending stairs, presumably a compensatory mechanism to reduce stress and therefore pain, familiarly known as the “quadriceps avoidance gait” (Brechtter & Powers, 2002). These authors speculated that forward inclination of the trunk to move the superincumbent center of mass over the knee was one mechanism by which knee joint moments were reduced, although they did not measure this motion.

Problem Statement

Given the frequency with which patellofemoral pain is observed and the disparate theories on how to manage it, clinicians have called for expanded research, particularly in the area of clinical studies, to better
understand the etiology and treatment of this malady (Henry, 1989; Grabiner, et al., 1994; Powers, 1998; Dye 1999; Powers & Mortenson, 1999; Powers, 2003). It is theoretically possible that hip abductor weakness influences femoral alignment and may contribute to PFP. (Ireland, Wilson, Ballantyne & Davis, 2003). Likewise, the role of the pelvis in forming the base from which the hip musculature maintains femoral alignment has been investigated in a two-subject \textit{case-study} experiment (Mascal, et al., 2003) but in no multi-subject investigational designs. As such, there is a need to investigate if differences exist in pelvic girdle alignment between groups of PFP and pain-free subjects that might lead to femoral alignment changes and consequent changes in knee biomechanics.

\textbf{Purpose}

In this study, individuals presenting with patellofemoral pain during descent of stairs underwent both kinematic and kinetic analyses of the knee, hip, and trunk, and kinematic analysis of the pelvis during descent of stairs, with the results compared to age-matched, pain-free subjects. These data were compared to previous work in this area to validate our measurement methodology, and to add to the body of knowledge on patellofemoral pain and the influence of biomechanical parameters from the pelvis and trunk.
Research Hypotheses

During descent of stairs subjects with patellofemoral pain would...

1. Descend stairs with lower peak and average knee flexion angles and moments compared to pain-free subjects.
2. Have greater knee valgus angles and moments compared to pain-free subjects.
3. Have greater hip adduction angles and moments compared to pain-free subjects.
4. Have greater peak and average pelvic girdle "obliquity" angles in the frontal plane compared to pain-free subjects.
5. Have greater trunk side-bending angles and moments compared to pain-free subjects.
6. Have greater trunk flexion angles and moments compared to pain-free subjects.

Significance of the Study

This study examines the kinematic and kinetic differences in women with and without patellofemoral pain when descending stairs, with particular attention paid to the relationship between motion of the trunk and pelvis and the kinematics and kinetics of the knee. Information from this study may help in the understanding of whether proximal stability differs in PFP patients and may assist clinicians in developing appropriate
interventions to address motion impairments separate from, but influencing, knee function.

Definitions:

**External joint moment.** Torque acting upon a joint due to ground reaction force from subject center of mass passing through joint. This value is equal and opposite to internal joint moment.

**Hip abductor strength.** A measure of subjects' ability to abduct the hip against gravity or investigator's manual resistance, as described in clinical manual muscle testing. Strength is graded on numeric scale, as described in methods. (Dvir, 1997)

**Internal joint moment.** Torque generated by muscles to oppose external joint moment. Value will be equal and opposite to external joint moment.

**Kinematics.** Objective measure of amplitude, velocity, and direction of limb and trunk motions without consideration of forces generating these motions. (Winter, 1990) For this investigation kinematics were measured using high-speed, infrared motion cameras, with data linked to a commercial software package for calculation of individual measures.

**Kinetics.** Estimation of moments acting on the knee, hip and trunk using inverse dynamics biomechanical formulas generated by force plate and kinematic data via a commercial software package. Moments were expressed as net external joint moments, or moments acting on the joint that must be balanced with internal (muscle action) moments.
Knee flexion motion. Greatest and average relative angles between thigh and shank during stance phase of gait during descent of stair, as determined by motion analysis methodology. (Salsich & Brechter, 2002).

Knee flexion moment. Peak and average external joint moment that would produce flexion about knee joint during stance phase of stair descent, as determined by inverse dynamic analysis of motion and force data.

Knee valgus/varus motion. Peak and average relative angle, in frontal plane, between thigh and shank, as determined by motion analysis methodology. Angles are considered valgus when apex is directed toward midline of body, while angle with apex away from midline is considered varus.

Knee valgus/varus moment. Peak and average external, frontal plane moment that would produce either valgus (tibial abduction) or varus (tibial adduction) motion about the knee.

Hip (femoral) adduction motion. Peak and average thigh adduction angle, relative to pelvis, during stance phase of stair descent, as determined by motion analysis methodology. (Mascal, et al., 2003)

Hip adduction/abduction moment. Peak and average external joint moment that would produce either adduction or abduction motion about hip joint during stance phase of stair descent, as determined by inverse dynamic analysis of motion and force data.
Joint stress. Joint reaction force divided by joint surface area (contact area) for a given joint (Powers, 2000).

Patellofemoral pain. Subject's report of specific or vague sense of anterior or medial knee discomfort that can be reproduced with specific activities, such as ascent or descent of stairs, running, jumping, kneeling, or prolonged sitting, as described in inclusion criteria under research design, subjects. (Greisamer & McConnell, 1998).

Pelvic girdle obliquity motion. Peak and average pelvic girdle motion in the frontal plane, horizontal to floor in laboratory coordinate system, as determined by motion analysis methodology. Motion in positive direction is considered obliquity, or list, toward stance side, while negative motion is considered obliquity or "list" toward contralateral stance limb. (Mascal, et al., 2003)

Trunk flexion motion. Trunk motion in sagittal plane, relative to horizontal, where positive measures are considered "flexion," and negative measures are considered "extension."

Trunk side-bending motion. Trunk motion in frontal plane, relative to horizontal, where motion toward the stance limb is considered ipsilateral, and motion away from the stance limb is considered contralateral.

Trunk flexion moment. Greatest and average external joint moment that would produce flexion about the lumbosacral joint during stance
phase of stair descent, as determined by inverse dynamic analysis of motion and force data.

**Trunk side-bending moment.** Greatest and average external joint moment that would produce side-bending about the lumbosacral joint during stance phase of stair descent, as determined by inverse dynamic analysis of motion and force data.

Theoretical Basis for Investigation

Despite the poorly understood etiology of patellofemoral pain, sufficient evidence exists to suggest that increased joint stress due to malalignment of the patella and femoral joint surfaces is a primary factor (Goodfellow, et al., 1976; Henry 1988; O'Neill, et al., 1992; Grabner, et al., 1994; Greitsamer & McConnell, 1998; Nati, Kannus, & Jarvinen, 1998; Juhn, 1999; Fulkerson, 2002). As such, researchers and clinicians have focused their interventions on realignment of the patella on the presumably stable femur.

More recently, however, attention is being paid to the position of the femur beneath the patella as possibly being the “malaligned” bony structure. Clinicians are often taught to observe the relative position of the patellae in the frontal plane to estimate femoral torsion, with excessive internal rotation of the femurs producing the so-called “squinting patellae” sign (Barovetz, et al., 1996; Juhn, 1999; Fulkerson, 2002). Others have advocated clinical estimation of relative femoral anteversion and
retroversion as an overall lower extremity biomechanical evaluation during examination of the patellofemoral pain patient (Gross, 1995; Jackson, 1996). Still others have advocated looking to the pelvis as a possible source of femoral malalignment, with anterior or posterior rotation of the innominate changing femoral alignment, ultimately producing patellofemoral symptoms (Jackson, 1996; Schramberger, 2002).

Similarly, weakness of the hip abductors (Ireland, et al., 2003) and pelvis and trunk musculature (Mascal, et al., 2003) has been cited as a possible contributing factors to femoral alignment.

One such category of those amenable to rehabilitation may be those with so-called “functional” anteverision, presumably caused by weak hip external rotators allowing a relative internal rotation of the femur (Kendall, McCreary, & Provance, 1993). Here the bony femoral alignment is near to normal, however muscular weakness produces functional malalignment. In fact, one study demonstrated a significant weakness in hip abductor strength in patients with patellofemoral pain compared to those without (Ireland, et al. 2003). More recently Chibulka and Threlkeld-Watkins (2005) reported a single-case study where “asymmetrical” hip rotation and hip abductor weakness were found in the ipsilateral knee of a PFP subject. Intervention aimed at strengthening the hip musculature and restoring normal hip rotation produced complete relief of patellar pain.

Finally, Tyler, Nicholas, Mullaney, and McHugh (2005) reported a series of thirty-five PFP patients who were treated with general hip stretching and
strengthening exercises (not just hip abductors) with a resulting improvement in 83% of cases. As such, clinicians have advocated strengthening of the hip musculature in the hopes of better aligning the femur beneath the patella (McConnell, 1998).

Similarly, researchers have noted increased internal rotation of the femur during landing from jumps in female athletes, speculating that this may predispose them to anterior cruciate ligament (ACL) injury (Griffin, 2000). Programs designed to strengthen the hip external rotators, among other activities, have markedly reduced the incidence of ACL injury in these athletes (Silvers, 2002).

Finally, investigators have looked at the pelvis as part of a total biomechanical assessment of persons with PFP and, in case study format, have shown differences in frontal plane alignment of the pelvis that were amenable to exercise intervention (Mascal, et al. 2003). However, no multiple subject studies have looked at pelvic girdle or trunk mechanics among those with patellofemoral pain.

Despite the increasing theoretical appreciation for proper femoral alignment, particularly as it is influenced by hip strength and pelvic girdle alignment, few investigational trials have looked at differences between those with and without patellofemoral pain, other than anecdotal testimonies (McConnell, 1986; McConnell, 2001) and case presentations (Mascal, et al., 2003).
CHAPTER II
REVIEW OF RELATED LITERATURE

Patellofemoral Pain and Stair Descent

Individuals with patellofemoral pain most often complain of symptoms following prolonged sitting ("movie goers" sign) and during activities requiring high quadriceps demand, such as ascent and descent of stairs, jumping, and running (Wilk, Escamilla, Fleisig, Barentine, Andrews, & Boyd, 1996; Neptune, Wright, & van den Bogert, 1999; Salsich et al., 2001; Taylor & Walker, 2001). Additionally, many FFP patient's chief complaint is retropatellar pain during descent of stairs while ascent of stairs is relatively pain-free, (Fulkerson, 2002) despite the higher knee moments found during stair ascent (Diedrich & Warren 1998; Salsich, et al., 2001; Costigan, et al., 2002). This phenomenon has previously been attributed to selective eccentric weakness of the quadriceps (Bennett & Stauber, 1986) although more recent theories suggest altered lower extremity alignment, particularly increased knee valgus, that may originate from proximal hip weakness leading to altered femoral alignment (increased adduction) (Powers, 2003). Following is a review of research looking at biomechanical changes of the lower extremities that may contribute to patellofemoral pain.
Knee Valgus Angle and Patellofemoral Pain

Varus and valgus deformities of the knee have long been associated with osteoarthritis change of the tibiofemoral joint and have more recently been linked to pain and arthritis of the patellofemoral joint. Elahi, Cahue, Felson, Engelman, & Sharma (2000) performed standing radiographs of older patients with patellofemoral pain and found a greater incidence of lateral patellofemoral osteoarthritis than medial and that lateral osteoarthritic changes were correlated with a valgus knee alignment.

In a follow-up study, Cahue, et al. (2004) looked at the 18-month progression of osteoarthritic (OA) changes of the patellofemoral (PF) joint and again found a high correlation (adjusted OR 1.64, 95% CI 1.01-2.66) between lateral patellofemoral changes and valgus knee angles. The authors suggested that; "Interventions that address the stress imposed by alignment on the patellofemoral compartments may delay PF OA progression and should be developed."

Finally, Fukui, Nakagawa, Murakami, Hiraoka, and Nakamura (2003) described a stress radiography technique in which subjects with suspected patellofemoral instability and normal subjects were radiographed with a mechanical valgus force applied to the knee and then compared to traditional Merchant views. Significant differences in patellofemoral congruence angles were seen between the groups for both
techniques, but with a significant correlation found between the stress x-rays and functional scores (measures of pain and functional ability) for the instability group.

Thus, clinical correlations have been made between knee valgus angle and patellofemoral pain and osteoarthritic change, leading some authors (Cahue, et al., 2003) to call for interventions aimed at addressing this bony alignment.

Femoral adduction angle and knee valgus angle

Increased knee valgus, particularly in women, has been attributed to increased femoral adduction angle (Horton & Hall, 1989). While often attributed to greater pelvis width in women, these authors attribute the corresponding increase in knee valgus to a greater hip width to femur length ratio compared to men. However, these measures were taken in static postures.

Looking at the knee during dynamic activity (running) Ferber, Davis and Williams (2003) found greater knee valgus angles in women compared to men during running that they correlated to greater peak hip adduction and internal rotation angles. The authors concluded "...further studies are needed to examine whether these differences in joint mechanics are related to the differences in injury patterns in males and females."
Fixed Femoral Alignment versus Hip Abductor Weakness

While femoral adduction angle has been linked to increased knee valgus, the cause or causes of increased femoral adduction must also be considered. In rare cases, a significant, fixed "coxa vara" deformity of the proximal femoral angle will produce a marked compensatory knee valgus deformity, ultimately requiring surgical correction (Shim, Kim, Mubarak, & Wenger, 1997). However, less significant coxa vara is also correlated with increased knee valgus (Hefti, 2000).

Separate from these fixed, bony deformities, increased femoral adduction may also be caused by weak hip abductors. During the stance phase of gait, including stair ascent and descent, the center of mass of the superincumbent body mass passes medial to the hip joint, causing an external hip adduction moment (Perry, 1992). This joint torque must be counterbalanced externally by the hip abductors to maintain equilibrium. Otherwise, weakness of the hip abductors can result in greater thigh adduction angle.

Finally, patellofemoral pain has a greater incidence in females than males (Fulkerson 2002), theoretically due to the fact that females tend to have weaker hip abductors (Griffin 2000, Lephart, Ferris, Riemann, Myers, & Fu, 2002; Silvers, 2002). Providing support for this theory, Ireland, et al. (2003) looked at hip abductor strength among women with patellofemoral pain. Comparing fifteen age-matched women with and without
patellofemoral pain, they found 26% less hip abductor strength among the painful group.

Pelvis Stability

Since the hip abductors originate from the lateral pelvis, stability of the pelvis has been theorized as a prerequisite for optimal lower extremity function (Sahrmann, 2002; Prentice, 2001). Including the management of patellofemoral pain (Greisamer & McConnell, 1998). However, only one group of researchers (Mascal, et al., 2003) have looked at pelvis motion among patellofemoral pain subjects. In reporting a two-subject case study, they demonstrated improved pelvis and femoral alignment, and reduced patellofemoral pain, following an exercise regimen aimed at the hip, pelvis, and trunk musculature. The authors concluded that; “Further research is indicated to better define the relationship between proximal muscle weakness and PFP, and to identify patients who will best respond to this treatment approach.”

Trunk Motion

Individuals experiencing patellofemoral pain during ambulation will often alter their gait in an attempt relive the discomfort. One strategy is to reduce the cadence or velocity of descent to lessen the overall force (moment) passing through the extensor mechanism (Brechter & Powers, 2002). Another common strategy is to reduce the demand of the quadriceps by employing what Perry (1995) termed the “quadriceps
avoidance gait." These individuals appear to incline the trunk to shift the center of mass in front of the knee to reduce the flexion moment that would otherwise be countered by the extensor mechanism. (Salsich, et al. 2001; Brechter & Powers 2002; Salsich, et al., 2002) Additionally, individuals with hip abductor weakness may side-bend the trunk toward the stance limb to reduce the hip adduction joint moment as a compensation strategy. Despite these clinical observations, no investigation has measured trunk inclination as part of their kinematic analyses. Therefore, this study will include measurement of trunk motion looking for differences between PFP and pain-free subjects.

**Literature Review Related to Methodology**

**Clinical Measurement of Hip Abductor Strength**

Standardized hip abduction strength has been established based on the work of Kendall, et al. (1993). This involves having subjects lie on their side, then attempt to raise the lower limb into abduction. Successful completion of full range of motion against gravity is graded as "Fair," or numerically, 3/5. The ability to resist mild to moderate manual resistance beyond gravitational force is considered "Good" or 4/5, while ability to resist greater manual force is considered "Normal," or 5/5. Inability to raise the limb against gravitational resistance is considered "Poor" or 2/5, and inability to move the limb with gravity eliminated is graded as "Trace" or 1/5 (Table 1).
Table 1.

Summary of hip abduction manual muscle testing grades. (Adapted from Kendall, McCreary and Provance, 1993).

<table>
<thead>
<tr>
<th>Result of manual muscle testing</th>
<th>Muscle strength grade</th>
</tr>
</thead>
<tbody>
<tr>
<td>The ability to resist moderate manual resistance beyond gravitational force</td>
<td>5</td>
</tr>
<tr>
<td>The ability to resist mild manual resistance beyond gravitational force</td>
<td>4</td>
</tr>
<tr>
<td>Successful completion of full range of motion against gravity, no manual resistance</td>
<td>3</td>
</tr>
<tr>
<td>Inability to raise the limb against gravitational resistance</td>
<td>2</td>
</tr>
<tr>
<td>Inability to move the limb with gravity eliminated</td>
<td>1</td>
</tr>
</tbody>
</table>
Using this method, many otherwise healthy young adults demonstrate hip abduction strength graded at 3/5, or less (Kendall, et al., 1993). This may be due to the long lever arm of the lower extremity which the hip abductors must work against, compared to the relatively short lever arm for other limb and limb segments used to judge other muscle groups (Dvir, 1997). Nonetheless, this measure of hip abductor strength is frequently used for clinical documentation purposes.

Variability of Kinetic and Kinematic Analysis of Stair Climbing

Several studies have examined knee kinematics and kinetics during stair ascent and descent in both normal (Asplund & Hall, 1995; Yu, Kienbacher, Growney, Johnson, & An, 1997; Costigan, et al., 2002) and persons with patellofemoral pain (Salsich et al., 2001; Brechter & Powers, 2002; Salsich, et al., 2002). In general, the reproducibility of measures in the sagittal plane (knee flexion and extension joint angles and moments) has been good, with similar normalized findings between studies, particularly when the measured step is preceded by a similar step, rather than a transitional stride from a level surface (Yu, et al., 1997). However, measures in the frontal plane (knee abduction and adduction joint angles and moments) and transverse plane (knee internal and external rotation angles) have been more variable (Yu, et al. 1997).
Fabrication of Laboratory Stairs and Influence on Gait

In their report on the effects of stair inclination and age on ground reaction force during ascent and descent of stairs, Stacoff, Kramers-deQuervain, Luder, List, and Stussi, (1998) summarized prior methodologies in fabricating stairs with force plates. They reported as few as two, and as many seven steps being used, with force plates embedded in the last or next to last step. Their study revealed fairly high variability and asymmetry among individuals when the step was configured with a steep (41-degree) incline compared to "flat" (20-degree) and standard (31-degree) inclines.

As reported by Yu, et al (1997) above, the first step when transitioning from a flat surface to stairs produces a varied gait compared to sequential steps. As such, it is important that a descending step precedes a data-collection step to reduce variability.

In this study, a three-step staircase will be fabricated to allow for a preceding step, however, the requirement to use an 8-inch step rise (see methodology) produces a 38-degree inclination, or a more varied "steep" inclination.

Reliability and Validity of Kinematic and Kinetic Data Collection Laboratory

At present, there are no standardized calibration protocols to determine the reliability and validity of kinematic and kinetic data collection procedures from one laboratory to another. This has brought about a call
for standard methods to test and validate different labs to ensure
uniformity in data collection procedures so comparisons between
researchers can be made (Selby, 2003). With this in mind, a commercially
available software package was created by Selby, et al. (Cal-Tester, C-
Motion, Inc, Gathersburg, MD) wherein motion and force data are
collected and independently compared to an externally generated "gold-
standard" for accuracy of data presentation. This is accomplished by
moving a rigid bar, with reflective markers of known dimension, in three
planes while directing a downward force into a live force plate with a fixed
center of pressure. In this way a rigid "vector" of force is generated which
is then compared to the internally generated force vector of the software
used for data collection, in this case Qualysis Track Manager (Qualysis,
Inc.; Gastonbury, CT). This device was employed prior to data collection
to demonstrate the accuracy of kinematic and kinetic measures in our
laboratory. Results of Cal-Tester testing and a photo of the device in use
are found in Appendixes H and I, respectively.
CHAPTER III

METHODS

Subject Selection and Screening

Ten women between the ages of 18 and 35 with primary complaint of anterior knee pain during descent of stairs were recruited for the investigation (Age = 22.0 ± 2.35 years; Height = 1.57 ± .11 m, mass = 64.72 ± 8.25 kg). Age, height, body mass, body mass index (BMI) patellofemoral pain (PPD) and level of activity were measured and recorded as described below and summarized in the following table.

Table 2.

Description of patellofemoral pain subjects.

<table>
<thead>
<tr>
<th>Subject</th>
<th>PPD Pain m</th>
<th>Height m</th>
<th>Mass kg</th>
<th>Activity</th>
<th>Hip MMT</th>
<th>Age</th>
<th>BMI</th>
</tr>
</thead>
<tbody>
<tr>
<td>8</td>
<td>3.8</td>
<td>1.6</td>
<td>59.5</td>
<td>2</td>
<td>5</td>
<td>22</td>
<td>21.7</td>
</tr>
<tr>
<td>10</td>
<td>3.8</td>
<td>1.57</td>
<td>53.2</td>
<td>3</td>
<td>4</td>
<td>24</td>
<td>21.5</td>
</tr>
<tr>
<td>14</td>
<td>4.5</td>
<td>1.7</td>
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<td>4</td>
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<td>23</td>
<td>24.9</td>
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<td>1.57</td>
<td>68.2</td>
<td>3</td>
<td>3</td>
<td>20</td>
<td>27.7</td>
</tr>
<tr>
<td>17</td>
<td>5.2</td>
<td>1.52</td>
<td>65.9</td>
<td>4</td>
<td>3</td>
<td>23</td>
<td>28.5</td>
</tr>
<tr>
<td>18</td>
<td>5.7</td>
<td>1.75</td>
<td>51</td>
<td>4</td>
<td>4</td>
<td>26</td>
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</tr>
<tr>
<td>23</td>
<td>4.6</td>
<td>1.6</td>
<td>72.7</td>
<td>4</td>
<td>4</td>
<td>19</td>
<td>22.4</td>
</tr>
<tr>
<td>26</td>
<td>3.5</td>
<td>1.78</td>
<td>70.5</td>
<td>4</td>
<td>3</td>
<td>20</td>
<td>22.2</td>
</tr>
</tbody>
</table>

SD 0.85 0.11 9.2% 1.01 0.71 2.35 3.56

M 4.06 1.57 64.72 3.44 3.07 22 23.44
A sample of 10 age-matched control subjects were then chosen (Age = 23.67 ± .71 years; Height = 1.64 ± .06 m; mass = 56.32 ± 4.25 kg). As with the patellofemoral pain subjects, age, height, body mass, body mass index (BMI) patellofemoral pain (PFP) and level of activity were measured and recorded as described below and summarized in the following table.

Table 3.

**Description of control subjects.**

<table>
<thead>
<tr>
<th>Subject</th>
<th>PFP</th>
<th>Pain</th>
<th>Height m</th>
<th>Mass kg</th>
<th>Activity</th>
<th>Hip MMT</th>
<th>Age</th>
<th>BMI</th>
</tr>
</thead>
<tbody>
<tr>
<td>11</td>
<td>0</td>
<td>0</td>
<td>1.57</td>
<td>54.5</td>
<td>1</td>
<td>3</td>
<td>24</td>
<td>22.2</td>
</tr>
<tr>
<td>15</td>
<td>0</td>
<td>0</td>
<td>1.65</td>
<td>58.2</td>
<td>3</td>
<td>4</td>
<td>23</td>
<td>21.4</td>
</tr>
<tr>
<td>16</td>
<td>0</td>
<td>0</td>
<td>1.66</td>
<td>56.8</td>
<td>2</td>
<td>5</td>
<td>24</td>
<td>20.1</td>
</tr>
<tr>
<td>21</td>
<td>0</td>
<td>0</td>
<td>1.55</td>
<td>52.3</td>
<td>1</td>
<td>5</td>
<td>25</td>
<td>21.8</td>
</tr>
<tr>
<td>22</td>
<td>0</td>
<td>0</td>
<td>1.75</td>
<td>63.6</td>
<td>3</td>
<td>5</td>
<td>24</td>
<td>20.8</td>
</tr>
<tr>
<td>25</td>
<td>0</td>
<td>0</td>
<td>1.6</td>
<td>59.1</td>
<td>2</td>
<td>4</td>
<td>23</td>
<td>23.1</td>
</tr>
<tr>
<td>27</td>
<td>0</td>
<td>0</td>
<td>1.65</td>
<td>57</td>
<td>2</td>
<td>4</td>
<td>24</td>
<td>20</td>
</tr>
<tr>
<td>29</td>
<td>0</td>
<td>0</td>
<td>1.68</td>
<td>56.8</td>
<td>2</td>
<td>5</td>
<td>23</td>
<td>30.1</td>
</tr>
<tr>
<td>30</td>
<td>0</td>
<td>0</td>
<td>1.6</td>
<td>48.6</td>
<td>1</td>
<td>4</td>
<td>23</td>
<td>19</td>
</tr>
<tr>
<td>31</td>
<td>0</td>
<td>0</td>
<td>1.68</td>
<td>53.5</td>
<td>2</td>
<td>4</td>
<td>22</td>
<td>20.2</td>
</tr>
</tbody>
</table>

SD: 0 0.06 4.25 0.78 0.71 0.70 1.28

M: 1.64 56.32 1.88 4.3 23.67 20.94
Table 4.

Summary of PFP versus Pain-free Subjects.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>PFP Subjects</th>
<th>Pain-free Subjects</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>PFP pain</td>
<td>4.06 ± .85</td>
<td>0.0 ± 0</td>
<td>p=.0001*</td>
</tr>
<tr>
<td>Height</td>
<td>1.67 ± .11</td>
<td>1.64 ± .06</td>
<td>p=.74</td>
</tr>
<tr>
<td>Body mass</td>
<td>64.7 ± 9.20</td>
<td>56.3 ± 4.25</td>
<td>p=.10</td>
</tr>
<tr>
<td>Activity Level</td>
<td>3.4 ± 1.01</td>
<td>1.89 ± .78</td>
<td>p=.01*</td>
</tr>
<tr>
<td>Hip MMT</td>
<td>3.7 ± .71</td>
<td>4.33 ± .71</td>
<td>p=.035*</td>
</tr>
<tr>
<td>Age</td>
<td>22 ± 2.40</td>
<td>23.7 ± .71</td>
<td>p=.70</td>
</tr>
<tr>
<td>BMI</td>
<td>23.4 ± 3.56</td>
<td>20.9 ± 1.3</td>
<td>p=.023*</td>
</tr>
</tbody>
</table>

(* significant difference p<.05)

Inclusion Criteria

Inclusion criteria were the following: 1) anterior knee pain during descent of stairs, and 2) any one or more of the following: a) anterior knee pain following prolonged sitting, b) anterior knee pain during ascent of stairs, c) anterior knee pain during kneeling, d) anterior knee pain during running, d) anterior knee pain during jumping, and e) anterior knee pain during isometric quadriceps contraction.
Exclusion Criteria

Exclusion criteria included any of the following: 1) any prior surgeries to the symptomatic knee, 2) any neurological condition affecting the lower extremities that would influence gait, 3) any ligamentous instability of the either knee that would influence gait, and 4) history of traumatic patellar dislocation.

Subject Screening

Prior to data collection, subjects were screened for eligibility and group assignment. They were fully informed of the nature and requirements of the study, and read and signed an informed consent form approved by the Seton Hall University IRB (Appendix C). They then completed a questionnaire regarding the nature of their present level of knee discomfort (Appendix D).

Next, they underwent a knee examination screening by the primary investigator to ensure compliance with both inclusion and exclusion criteria, with results recorded on a form (Appendix E).

Subject's height and mass were determined and their body mass index (body mass/stature) was calculated.

Additionally, subjects' self-reported activity level was determined according to the following scale developed by the author, for descriptive purposes only:
1. No physical activity, other than activities of daily living.
2. Occasional physical activity, but no regimented activity program
3. Light to moderate physical activity at least 2-3 times per week.
4. Regular, vigorous activity 3-5 times per week.
5. Vigorous or competitive physical activity 5 or more times per week.

Knee discomfort during stair descent on the day of testing was measured by having subjects descend a standard stair case adjacent to the motion lab, then recording their level of discomfort on a 10-centimeter visual analog scale (Appendix F). Their mark was later measured by the investigator from the left terminus and recorded in centimeters on the form.

Hip abduction strength was assessed using a standardized manual muscle testing procedure (Kendall, et al., 1993) with the result recorded on a data collection form (Appendix G).

Instrumentation

Three-dimensional (3-D) video-image data were initially collected using four then five (a fifth camera was obtained during data collection) Qualysis MCU 1000 high-speed, video cameras (Qualysis, Inc.; Glastonbury, CT) at 120 Hz with data stored and later analyzed using Qualysis Track Manager (QTM) software (Version 1.6.0.171, Qualysis, Inc.; Glastonbury, CT). Force data were collected via a Bertec model
4060A force plate (Bertec, Inc; Columbus, OH) embedded within the staircase at a sampling rate of 1,200 Hz per channel and synchronized with video data using QTM software. All software applications were run on a Dell desktop computer with an 80 gigabyte hard-drive, 1 GHz processor, and 256K random access memory.

As noted in Chapter Three, a commercially available software package (Cal-Tester, C-Motion, Inc, Gathersburg, MD) was employed to ensure that video-image data and force plate data were properly synchronized to produce valid resultant force vectors that would later be used for inverse dynamics joint moment calculations.

Included in Appendix H are results of Cal-Tester data that measured the accuracy of the resultant force vector from the center of pressure reported by the force plate compared to a known force vector generated by a rigid arm "gold-standard" described earlier. This report revealed a small difference (1.3 ± 0.7 degrees) between the force vector generated by Qualysis software and that generated by the rigid body "gold-standard" generated with Cal-Tester. While standardized measures of error have not been established, the chief developer of Cal-Tester software reviewed this report (personal communication, Scott Selbie, May 24, 2004) and indicated that the values were within acceptable ranges (less than 2.0 degrees) to be considered valid.
Video cameras were positioned on either side and in front of the staircase, offering good frontal and sagittal plane views of subjects, but no posterior views. (Figure 7).

Figure 7. Floor plan of 5-camera data collection lab (dimensions not to scale).

Staircase

A staircase was made by stacking two 4-inch platforms, one with an embedded force plate, to create an initial 8-inch step. A wooden flight of two, 8-inch steps was fabricated and rested upon the stacked platforms, producing a three-step staircase. The standard rise of steps is between 7 and 8 inches, making these steps at the higher range of standard construction. (Figure 8).
Figure 8. Staircase with embedded force platform. Each step measures 8-inches of vertical dimension.
Procedure: Calibration

Prior to each data collection session the instruments were calibrated according to manufacturer's directions. This involved passing a wand with two reflective markers of known distance throughout the data collection field, which also contained a static, 4-marker, L-shaped frame of known marker distances resting atop the force plate. When the camera system "captured" the marker distances with standard error of 1.5 mm, or less, compared to the known marker distances, the camera system was considered to be calibrated. This data collection also served to define the (global) laboratory coordinate system, and to produce a virtual laboratory perspective for future animation purposes.

Marker System

Nineteen-millimeter retro-reflective markers were placed over bony landmarks of the trunk, pelvic girdle, hip, knee, and ankle using adhesive Velcro patches or double-sided cellophane tape according to NIH guidelines (Figure 9). These markers served to establish location of joint centers, and length of bony segments, to be used in skeletal modeling during data analysis. Two markers were used to establish each joint center, for a total of 10 markers. These were affixed at the ASIS of the ilia for the pelvis, over the greater trochanters for the hip, over the medial and lateral tibial condyles for the knees, and over the medial and lateral malleoli for the ankle. Next, three additional retro-reflective markers were
similarly affixed to the trunk, pelvis and foot, while pre-formed plastic shells affixed with reflective markers were strapped to the thigh and shank in a comfortable manner using flexible adhesive tape. These markers served as segmental "motion" markers used to establish movement of bony segments during motion data collection. A total of 15 markers were used for this purpose. Subjects then descended stairs as needed to assure unencumbered freedom of movement with the markers in place, with correction of marker adherence and strapping made as needed.

Subjects then underwent a 2-second stationary 3-D video image recording of marker placements by standing at ease within the camera field of focus. These data were then saved using a code involving their subject number, but no identifying name, for later use in software modeling of the bony skeleton and biomechanical motion. Each of the trunk, pelvic girdle, hip knee and ankle joint markers were then removed, leaving just the segmental motion markers required for collection of motion data.
Figure 9. Example of marker placements of joint center landmarks and motion segments.
To avoid the possible effects of footwear on gait when descending stairs, all subjects underwent data collection in their stocking feet. Subjects were familiarized with the test procedure, including practicing descent of three, 8-inch steps. They were instructed to begin descent of stairs with the uninvolved limb, then continue descent with the involved limb such that it struck a pre-aligned force plate allowing analog force data measures. A final step with the uninvolved limb and continued walking on level ground completed the task. Subjects were allowed to repeat the procedure as often as necessary to gain confidence and consistency, but were cautioned to not aggravate their symptomatic knee.

Data collection trials for stair descent began with subjects standing at ease atop the staircase. Upon a command by the investigator, subjects began descent of stairs while the motion cameras and force plate were activated to record data. Trials were considered successful if the subjects stepped on the force plate area during descent. Data were again saved using an anonymous code. Five trials per subject were conducted. Subjects in the pain-free group underwent identical testing, using markers on either the right or left lower extremity in order to match the number of right and left symptomatic knees from the patellofemoral pain group. In all cases they responded at the far left terminus ("No pain") of the visual analog scale of knee pain during descent of stairs, or were otherwise excluded from data collection that day.
Kinematic and Kinetic Data Analysis

Kinematic data were analyzed by first identifying markers according to a standardized naming system in Qualysis Track Manager. This software also has the capability of “interpolating” trajectories of markers where markers might be briefly “lost” when they cannot be seen by two or more cameras. Based on mathematical algorithms, a best-fit trajectory is then interpolated. In all cases, interpolated trajectories were visually inspected for accuracy, and manually adjusted by the investigator. In cases where accurate trajectories could not be made, the trial was rejected.

Both kinematic and kinetic data were then saved and exported to Visual 3D software (C-Motion, Inc., Rockville, MD) via C3D file format, an industry-standardized file format containing identified marker (motion) data, force (kinetic) data, force plate matrices, and related data parameters in a single file format. C3D software is a public domain software package made available by Motion Lab Systems (Motion Lab Systems, Inc., Baton Rouge, LA).

In Visual 3D kinematic data were filtered with a second-order Butterworth bi-directional filter (resulting in a fourth order filter) with a low-pass cut-off of 6 Hz, and kinetic (analog) force data (ground reaction force) were loss-pass filtered at 20 Hz. In order to address a tendency for the analog force plate signal to "drift" during the first 0.5 seconds of data collection, a device within Visual 3D software was used wherein the
analog signal was “zeroed” at time interval 1.50 to 1.60 seconds after start of data collection. Therefore, when subjects struck the force plate at between 2.0 and 3.0 seconds the “drift” would have been eliminated. Visual inspection of analog data tracing demonstrated that this technique was successful in removing the errant data.

To ensure proper orientation of the force plate within the data collection field, and importation of accurate force plate parameters, a programming “script” containing previously recorded force plate dimensions was entered into the Visual 3D computations. Next, timing of “events” to begin and end reporting of kinematic and kinetic data, including calculation of joint moments via inverse dynamic computations, was achieved by using force plate thresholds of pressure (5 Newtons in the “Z” direction,) producing “on-off” instants of time on the force plate. Therefore, all data reports are for time spent stepping on the force plate only.

Modeling

For joint motions, a link-model-based format was used in which joint motions were determined by plotting the motion of rigid distal and proximal segments around pre-determined (via static data capture) joint centers. Angles are described according to a local coordinate system, with motion occurring around three joint axes where sagittal plane motion (flexion and extension) occurs about an “X” axis, frontal plane motion...
(abduction and adduction, and pelvic obliquity) occurs about a "Y" axis, and transverse plane motions (rotation) occur about a "Z" axis (Figure 10).

![Diagram of coordinate systems](image)

**Figure 10.** Segmental and laboratory coordinate system. (Adapted from C-Motion, Gaithersburg, MD)

To determine joint centers and segment lengths, the aforementioned joint markers were recorded during the two-second static data collection. For example, for the knee, medial and lateral markers over the distal femoral condyles served to determine the knee joint center, while medial and lateral malleolus markers determined the distal (shank)
segment, and a greater trochanter marker determined the proximal (thigh) segment. The segments were further defined by calculation of anthropometric dimensions according to height and weight data for individual subjects being inputted under subject data entry. This anthropometric data is also used in calculation of joint moments, below.

Changes in joint motion for the lower extremities were expressed in degrees according to either a local coordinate system (one segment in relation to another,) as used for the knee and hip, or according to a global coordinate system (one segment in relation to laboratory coordinates,) used for the trunk and pelvis. For motion in the sagittal plane (flexion and extension of the hip and knee) zero degrees represented a straight line between segments (no motion of the joint,) with each positive measure of degrees of “bending” (approximation of segment ends) represented as flexion, and negative degrees represented as extension. For motion in the frontal plane, a straight line between segments represented zero degrees of motion, with motion of the hip away from the midline of the body described as abduction, and motion toward the midline described as adduction (all measures positive). Likewise for the knee, no motion in the frontal plane between segments was considered zero degrees, with motion of the distal segment toward the midline described as varus, and motion away from the midline described as valgus.

For motions of the pelvis in the frontal plane, a parallel (horizontal) position to the laboratory floor (global coordinate system) was considered
zero degrees, with motion ("drop") toward the stance limb considered ipsilateral obliquity and were expressed as a positive number, while motion toward the opposite (swing) limb was considered contralateral obliquity and was expressed as a negative number.

While modeling of the lower extremities has been described elsewhere (Kepple & Arnold, 1994) modeling of the trunk is not often described. In this study the trunk was defined with static markers placed over the acromioclavicular joints to define superior length and width dimensions, and with markers placed over the iliac crests to define inferior length and width dimensions. As such, the abdomen and thorax are treated as a rigid segment, although in reality the abdomen and trunk may have independent motions. Three motion markers were affixed to the superior chest (pectoral and mid-sternal region) to track motion of the trunk during stair descent. Motion of the trunk may be described with reference to the pelvis (lumbopelvic joint) or with reference to the laboratory coordinate system (LCS). In this study, trunk motion is described with reference to the LCS since movement of the pelvis would confuse appreciation of whether the motion occurred at the trunk, pelvis, or both (Figure 11). Forward inclination of the trunk with respect to the LCS was considered trunk flexion; backward inclination was considered trunk extension; and sideward inclination in the frontal plane was considered side-bending of the trunk.
Calculation of trunk moments were similarly made using inverse dynamics equations and link segment models. However, the “link” made by the proximal end of the distal segment (hip joint) and distal end of the proximal segment (lumbosacral joint) have an intervening gap (ileum). This gap is sufficiently large to create error in moments calculated for the trunk and should be interpreted with caution (personal communication, Scott Selbie, April 11, 2006.).

Figure 11. Local (laboratory, LCS) coordinate system and segmental (lumbopelvic, SCS) coordinate system for modeling of trunk.
Joint Moments

Joint moment calculation was accomplished by employing link-model-based formulas which included ground reaction force and subject anthropometric data to produce inverse dynamic calculations for external joint moment reports. Visual 3D uses an algorithm that produces net internal joint moments, that are easily converted to external moments using a -1.0 multiplier.

Reports

Visual 3D allows custom report writing to display data. For this investigation, two general categories of reporting were used; 1) joint motions in each of three planes, including the knee, hip, and trunk, and in a single plane (frontal) for the pelvis, and 2) joint moments in each of three planes for the knee, hip and trunk.

Independent Variables:

1. Group (Patellofemoral pain versus pain-free controls).

Dependent variables of interest:

1. Peak and average knee flexion angles.
2. Peak and average knee valgus angles.
3. Peak and average hip adduction angles.
4. Peak and average pelvic obliquity angles.
5. Peak and average trunk flexion angles.
6. Peak and average trunk side-bending angles.
7. Peak and average knee flexion moment.
8. Peak and average knee valgus moment.
9. Peak and average hip adduction moment.
10. Peak and average trunk flexion moment.
11. Peak and average trunk side-bending moment.

Data Analysis; Data Selection

When subjects successfully completed a data collection trial (see Procedures, below) their data were stored and analyzed as described above. A total of up to five trials were recorded, analyzed and graphically reported such that all five trials could be superimposed and visually inspected for each of the independent variables cited above. To address intra-subject variability, “outlying” trials (trials with standard deviation greater than 1.9) were deleted (Figure 12.) This would occur when subjects took a “miss-step” on the stair, producing aberrant kinematic or kinetic tracings. Similarly, trials in which kinematic data were incomplete due to markers being out of camera view (each marker must be seen by at least two cameras to produce three-dimensional data) and were incompletely interpolated in Visual 3D (Figure 13). This was a fairly common occurrence when using a four-camera system, but infrequent when using five cameras.
Figure 12. Graph depicting trial with outlying data selected for deletion.
Figure 13: Graph depicting trial with incomplete data selected for deletion.
Once three trials for each subject were selected, the mean values for the entire curve for each kinematic or kinetic dependent variable were determined in Visual 3D (Figure 12). Mean data for each subject were then exported to Excel software via P2D file format. Kinematic data (joint angles) were saved as "raw" values, while kinetic data (joint moments) were normalized to subjects' body mass so comparison could be made to subjects of differing size. In Excel, peak and average values for each of the dependent variables noted above were calculated. Since P2D file format records individual data points in a temporal fashion "average" scores can be considered "area under the graph" averages, with angle or moment represented on the Y-axis and time represented on the X-axis. As such, subjects with similar minimum and peak joint angles or moments, but differing times spent at each angle or moment, would have different averages. Peak and average values were then exported to SPSS (version 12.0) for statistical analyses.

In SPSS the peak and average values for all subjects in each group (n=10) were statistically analyzed for inter-subject variability and differences between groups. Figure 14 summarizes the flow of data collection.
Figure 14. Data flow chart.

Static and dynamic motion captured via Qualysis 3-D video cameras.

Ground reaction force (GRF) recorded with Bertec force plate.

Markers identified and trajectories interpolated; kinematic and analog data converted to C3D file format in Qualysis Track Manager.

Analog (GRF) signal amplified.

Motion and Force data exported to Visual 3D. Joint angles and moments calculated. Results exported via P2D format.

Average and peak joint angles and moments calculated in Excel.

Mann-Whitney U statistical analyses applied in SPSS, version 12.0.
Descriptive Statistics

Descriptive statistics were used to describe and compare the two groups of subjects with means and standard deviations provided for characteristics such as age, activity level, height, body mass and index, knee pain level, and hip abduction strength. (Tables 3 and 4.)

Inferential Statistics

Since a relatively small number of subjects (n=10 for each group) were recruited, non-parametric statistical analyses were likewise used to analyze kinematic and kinetic data. Specifically, the Mann-Whitney U test was employed, with significance set at 0.05, a priori. In cases where “direction” of differences between groups was predicted a one-tailed test for significance was employed, while with measures of unknown direction of differences a two-tailed test for significance was used. All statistical computations were made using SPSS software (version 12.0).

Power Analysis

Previous researchers have found significant differences in knee kinematic and kinetic values with as few as 10 subjects in each of the symptomatic and control groups (Powers & Landel, 1997, Salsich & Brechter, 2001). To ensure sufficient numbers of subjects in this study a power analysis was conducted when 5 patellofemoral and 7 control subjects had completed data collection. This was done using Power
Calculator software, available for use online via the Internet. (UCLA Department of Statistics, 2005) Using the "non-parametric" option, significance level (Alpha) was set at 0.05 and effect size at 0.8 (Beta .20.) it was found that between 8 and 14 subjects would be needed to show significance, depending on the dependent variable chosen. As such, a total of ten subjects underwent data collection, followed by a trial run of non-parametric statistics. Significant differences were then seen in each of the key dependent variables of interest. Power analyses of those variables shown to not be significant revealed subject requirements of 20 to 50, or more, to demonstrate significance. Therefore, data collection was deemed to be adequate at n=10 for each group.
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Table 5. Power analysis table for kinematics, with Alpha set at .05, and Beta set at .20. N equals number required for significance.
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Table 6. Power analysis table for kinetics, with Alpha set at .05, and Beta set at .20. N equals number required for significance.
CHAPTER IV
RESULTS

Introduction

Results are presented for kinematic variables and kinetic variables in the order in which they were presented in the list of dependent variables. An additional statistical measure correlating peak pelvic obliquity angle and peak knee valgus moment is reported.

Kinematic Variables

Of twelve kinematic variables studied, four showed significant \((p<.05)\) differences. These included peak and average knee flexion angle, and peak and average pelvic obliquity angle in the frontal plane.

Kinematic variables found to be not significant were peak and average knee valgus angle, peak and average thigh adduction angle, peak and average trunk flexion angle, and peak and average trunk side-bending angle. Below are the specific findings for each of these measures.

*Knee flexion angle.* A significant difference was found for both peak \((p=.002)\) and average knee flexion angle \((p=.001)\) with patellofemoral pain subjects averaging lower peak flexion angles \((M = 80.0, \ SD \pm 11.1 \ degrees)\) compared to pain-free subjects \((M = 90.8, \ SD \pm 5.1\)
degrees.) Likewise, average knee flexion angles were lower for patellofemoral pain subjects (M 34.4, SD ± 2.8 degrees) compared to pain-free subjects (M 41.6, SD ± 4.2 degrees.)

![Graph showing joint angles for control and PFP groups](image)

**Figure 15.** Peak knee flexion angle (+SD) for control (n=10) and PFP subjects (n=10.) (U = 14.0, z = -2.72, p=.007)
Figure 16. Average knee flexion angle (+SD) for Control (n=10) and PFP subjects (n=10.) (U = 4.0, z = -3.48, p = .001.)
Figure 17. Knee flexion angle for control (n=10) versus PFP subjects (n=10.)
Knee valgus/varus angle. No significant differences were seen for peak knee valgus angle (p=.17) and average knee valgus angle (p=.85) between groups, although patellofemoral subjects tended to have greater peak valgus angles (M 9.5, SD ± 4.8 degrees) compared to pain-free subjects (M 4.9, SD ± 8.3 degrees).

Hip adduction angle. Patellofemoral pain subjects had, on average, greater peak (M 8.3, SD ± 2.6 degrees) and average (M 3.4, SD ± 2.7 degrees) hip adduction angles compared to pain-free subjects (M 1.5, SD ± 7.9 degrees; M 1.2, SD ± 3.9 degrees, respectively.) However, these differences were not statistically significant (peak hip adduction angle p=.10; hip average adduction angle p=.40.)

Pelvic obliquity angle. A significant difference was found between groups for both peak (p=.0001) and average (p=.0001) pelvic girdle obliquity (obliquity toward the stance limb, expressed as a positive number). Patellofemoral pain subjects had greater peak (M 5.0 SD ± 1.5 degrees) and average (M 2.1, SD ± 1.1 degrees) pelvis obliquity angles compared to pain-free subjects (peak -.04, SD ± 3.5 degrees; average -0.7, SD ± 1.4 degrees) where positive numbers represent obliquity toward the stance limb, while negative numbers represent obliquity opposite the stance limb. All pelvic girdle joint angles are with respect to the horizontal, laboratory coordinate system.
Figure 18. Peak pelvic obliquity angle (+SD) for control (n=10) and PFP (n=10) subjects. ($U = 0.0$, $z = -3.79$, $p = .001$)
Figure 19. Average pelvic obliquity (+SD) angle for control (n=10) and PFP (n=10) subjects. (U = 3.5, z = -3.52, p=.001)
Figure 20. Control (n=10) versus PFP (n=10) subjects for pelvis obliquity angle.

**Trunk flexion angle.** Since it was hypothesized that trunk flexion angle would be increased with PFP subjects, a one-tailed statistical analysis was employed. No significant differences (p=.35) were seen for peak trunk flexion angle between control subjects (M 11.24, SD 3.44) and PFP subjects (M 7.68, SD 3.50.) Likewise, no differences (p=.25) were seen for average trunk flexion angle between control subjects (M 8.01, SD 2.99) and PFP subjects (M 3.82, SD 3.07.)
Trunk side-bending angle. A one-tailed statistical analysis was applied for trunk side-bending motion. No significant difference for peak side-bending angle ($p=.45$) between control subjects ($M = 0.24$, $SD = 2.84$) and PFP subjects ($M = 0.16$, $SD = 7.18$). Likewise, no differences ($p=.35$) were seen for average trunk side-bending angle between control subjects ($M = -0.05$, $SD = 1.56$) and PFP subjects ($M = -0.96$, $SD = 4.51$).

Kinetic Variables

Of the joint moments calculated, six revealed significant differences. These included peak and average knee flexion moment, peak and average knee valgus moments, and peak and average hip adduction moments. Joint moments without significant differences were peak and average trunk flexion moment, and peak and average trunk side-bending moment.

Knee flexion moments. A significant difference was found between groups for both peak ($p = .061$) and average ($p = .001$) knee flexion moments. In both cases, patellofemoral pain subjects had reduced knee flexion moments, with peak moments of .94 Nm/kg ($SD = .11$ Nm/kg) compared to 1.20 Nm/kg ($SD = .16$ Nm/kg) for pain-free subjects. Average knee flexion moments were .37 Nm/kg ($SD = .06$ Nm/kg) for SD patellofemoral subjects and .56 Nm/kg ($SD = .11$ Nm/kg) for pain-free subjects.
Figure 21. Peak knee flexion moment (+SD) for control (n=10) and PFP subjects (n=10). (U = 7.5, z = -3.21, p = .001)
Figure 22. Average knee flexion moment (+SD) for control (n=10) and PFP subjects (n=10). \( (U = 7.5, z = -3.21, p = .001) \)
**Knee Flexion (⁺) Moment**

![Graph showing knee flexion moment comparison between PFP and control subjects.]

**Figure 23.** Control versus PFP subjects for knee flexion moment.

**Knee valgus/varus moments.** A significant difference was also seen for peak (p=.006) knee valgus moment and average (p=.002) knee valgus moment. Patellofemoral pain subjects had greater peak (.35 SD ± .36 Nm/kg) and average (.18 ± SD .14 Nm/kg) knee valgus moments compared to pain-free subjects, (peak -.13 SD ± .37 Nm/kg; average -.07 SD ± .15 Nm/kg).
Figure 24. Peak knee valgus/varus moment (+SD) for control (n=10) and PFP subjects (n=10). (U = 13.50, z = -2.76, p = .006.)
Figure 25. Average knee valgus/varus moment (+SD) for control (n=10) and PFP subjects (n=10). (U = 8.5, z = 3.16, p=.002.)
Figure 26. Graph of control (n=10) and PFP (n=10) peak knee varus/valgus moment.
Hip adduction moments. Significant differences were also seen for peak (p=.006) and average (p=.004) hip adduction moments, with patellofemoral subjects having greater moments (peak .65 SD \pm .21 Nm/kg; average .29, SD \pm .07 Nm/kg) than pain-free subjects (peak .40 SD \pm .06 Nm/kg; average .19, SD \pm .10 Nm/kg.)
Figure 27. Peak hip adduction moment (±SD) for control (n=10) and FFP subjects (n=10). (U = 14.0, z = -2.72, p=0.006)
Figure 26. Average hip adduction moment (+SD) for control (g=10) and PFP subjects (g=10). \(U = 12.50, z = -2.84, p = .004\)
Figure 29. Control (n=10) versus PFP (n=10) hip abduction/adduction moment.
Trunk flexion moments. No significant differences were seen for peak \((p=.68)\) and average \((p=.62)\) trunk flexion moments. Patellofemoral subjects had peak moments of \(.54 \text{ Nm/kg (SD } \pm .07)\) and average moments of \(.19 \text{ Nm/kg (SD } \pm .06)\) compared to peak moments of \(.56 \text{ Nm/kg (SD } \pm .13)\) and average moments of \(.21 \text{ Nm/kg (SD } \pm .07)\) for pain-free subjects.

Trunk side-bending moments. No significant differences were seen between groups for peak \((p=.52)\) and average \((p=.90)\) trunk side-bending moments, with patellofemoral pain subjects having peak \(.29 \text{ Nm/kg (SD } \pm .05)\) and average \(.10 \text{ Nm/kg (SD } \pm .03)\) moments, compared to pain-free subjects with peak \(.32 \text{ Nm/kg (SD } \pm .12)\) and average \(.09 \text{ Nm/kg (SD } \pm .08)\) moments.

Additional Results

Since a primary interest of this investigation is the influence of the pelvis on knee alignment, a correlation statistic was applied between peak pelvic obliquity angle and peak knee valgus moment, both of which were significantly greater in PFP subjects. In comparing peak pelvic obliquity angle and peak knee valgus moment for all subjects \((N=20)\), a significant, positive correlation \(r = .437, p = .027\) was found between these variables, with greater pelvic obliquity toward the stance side correlating with greater knee valgus moment (Table 7 and Figure 30).
Table 7.

Pearson product-moment coefficient of correlation between peak valgus/varus moment and peak pelvis obliquity angle.

<table>
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<th>Pearson Corr.</th>
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<td>20</td>
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<tr>
<td>Peak Pelvis Obliquity Angle</td>
<td>.437</td>
<td>.027</td>
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</table>

Peak Pelv Od Ang Pearson Corr. (r-value) | .437 |
Sig. (1-tailed) | .027 |
N | 20  |
Figure 30. Scatter plot of Peak Pelvic Obliquity Angle and Peak Knee Valgus/Varus Moment.

Additionally, highlighting the PFP patients (Figure 30) reveals this group to have the greatest pelvic obliquity angles correlating with the greatest knee valgus moments.

Following is a summary (Table 8) of significant kinematic and kinetic findings.
Table 8.

Summary of significant findings.

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<td>Cont. 41.6 ± 4.2</td>
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<td>Cont. -0.7 ± 1.4</td>
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<td>PFP 90.8 ± 5.1</td>
<td>PFP 34.4 ± 2.8</td>
<td>PFP 5.0 ± 1.5</td>
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<table>
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<th>Avg. Knee Flexion Moment (Nm/kg)</th>
<th>Peak Knee Valgus Moment (Nm/kg)</th>
<th>Avg. Knee Valgus Moment (Nm/kg)</th>
<th>Peak Hip Adduction Moment (Nm/kg)</th>
<th>Avg. Hip Adduction Moment (Nm/kg)</th>
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<td>Cont. 5.5 ± 0.11</td>
<td>Cont. 3.5 ± 0.36</td>
<td>Cont. 18 ± 0.37</td>
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<td>PFP 0.94 ± 0.11</td>
<td>PFP 3.7 ± 0.06</td>
<td>PFP -0.13 ± 0.37</td>
<td>PFP -0.07 ± 0.15</td>
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<td>p = 0.002</td>
<td>p = 0.096</td>
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</table>

Summary Graphs

To better appreciate the interdependence of joints in the biomechanical chain, kinematic graphs are arranged in top-down sequence (Figure 31), and both kinematic and kinetic graphs are presented in Figure 32. This will be further developed in Discussion section.
Figure 31. Top-down flow chart of joint angles.
Figure 32. Top-down flow chart of kinematic and kinetic graphs.
CHAPTER V
DISCUSSION

Introduction

The purpose of this study was to determine if differences exist between subjects with and without patellofemoral pain with respect to lower extremity, pelvis, and trunk kinematics and kinetics during stair descent and to better understand the cause or causes of the differences seen in the knee mechanics of PFP individuals. The initial hypotheses of less knee flexion angle and knee flexion moment in PFP subjects was supported and corroborated the findings of previous researchers (Salsich, et al., 2002).

Additional new findings were greater hip adduction moment and greater knee valgus moment among PFP subjects. Corresponding increases in hip adduction and knee valgus angles for PFP subjects, however, were not found, although a trend toward greater adduction and valgus was seen for this group. An additional significant finding, corroborating the work of others (Mascal, et al., 2003), was the observation of greater pelvis obliquity angle among PFP subjects, while control subjects demonstrated a relatively neutral pelvis. However, where Mascal et al. found increased contralateral pelvis obliquity, this study found significant ipsilateral pelvis obliquity.

Adding to the body of literature in this area was the finding of a positive correlation between knee valgus moment and pelvis obliquity angle among
all subjects, with PFP subjects having the greatest knee valgus moments and greatest ipsilateral pelvis obliquity angles. This association, while not suggesting a cause and effect, does lend some credibility to the "top down" theory of abnormal knee mechanics having a contribution from the hip and pelvis.

Hypotheses not supported were the expectations that trunk flexion and side-bending angles and moments would be different for PFP subjects as an explanation for the reduced knee moments consistently seen in this population. Possible explanations for these findings are discussed below.

For the remainder of this section, a review of the subject characteristics and each of the significant findings reported in Results, including a discussion linking kinematic and kinetic findings, will be shown and compared to the existing literature. A final Conclusion chapter will summarize the chief findings and suggest clinical implications.

Subjects

Ten subjects with patellofemoral pain (PFP) were recruited, and then matched with pain-free subjects of similar age. However, these differences were seen between the groups. First, PFP subjects were somewhat more active than their pain-free counterparts. This is similar to the findings of other investigators (Devereaux & Lachmann, 1984; Henry, 1983; Juhn, 1399; Steinkamp, et al., 1993; Powers, 1998; Malone, et al., 2002) where PFP patients were most often found among active, sports participants. A
common interpretation of this finding is that the subtle biomechanical
differences seen among PFP populations only result in painful conditions
when the knee is subjected to repeated or high stress activities (Powers,
2003), while those who are less active might be pain-free, although they too
may have biomechanical faults.

A second difference between groups was the finding that PFP
subjects, on average, had weaker hip abductors than pain-free subjects.
This finding is similar to that of Ireland, et al. (2003) where PFP subjects
had significantly weaker hip abductors, as measured by handheld
dynamometers. It should be noted, however, that in the current study a
clinically based assessment tool (traditional manual muscle testing) was
used to determine hip abductor strength as a simple descriptor of subjects,
rather than as a dependent variable for statistical analysis. As such, direct
comparison of these works should be approached with caution.

Finally, PFP subjects had a greater body mass index (BMI) than did
pain-free subjects. This could be accounted for by the fact that the PFP
subjects were active, often athletic individuals, some of whom were large
framed and heavily muscled. None could be described as obese, however,
so BMI alone was not considered a confounding variable to the study.

General Kinematic Discussion

Knee flexion angle. Both peak and average knee flexion angles were
reduced in PFP subjects compared to pain-free subjects. This is consistent
with the findings of other investigators looking at kinematics during stair descent under both normal conditions (no intervention) (Salsich, et al., 2001) and prior to application of adhesive tape to the patellofemoral joint (Salsich, et al., 2002). This observation is partly explained by the fact that with increasing knee flexion greater joint reaction force, and more importantly greater joint stress (Brecht, & Powers, 2002) is produced, often leading to patellofemoral pain. Those with pain will naturally limit knee bending motion as a strategy to avoid pain when descending stairs.

**Knee valgus angle.** Both peak and average knee valgus angles were increased in PFP subjects compared to pain-free subjects, although the differences were not statistically significant (p = 0.15). One of the principle hypotheses of this study was that PFP subjects would have increased knee valgus angles, thought to be a source of patellofemoral malalignment (Cahue, et al., 2004, Etahi, et al., 2000). While a trend was seen toward greater valgus angle in PFP subjects’ angles (M 9.5, SD ± 4.8 degrees) compared to pain-free subjects (M 4.9, SD ± 3.3 degrees) the high inter-subject variance, particularly among the pain-free subjects, made the differences not significant.

A possible explanation for the high within groups variance noted, and the observed high intra-subject variability between trials, is the relatively high (8-inch, 38-degree angle) step height used in the laboratory staircase for this study. As discussed in Methods above, Stacof, et al. (1998) reported that steep staircases (up to 41 degrees) produce increased
variability between trials for the same individual and may have added to the variability of measures in this study. When observing subjects descend stairs, subtle changes in their normal gait were observed, such as apparent widening of the base of support, slowing their cadence, and, initially, looking down at the staircase when descending. However, for all subjects, sufficient practice trials were permitted to acclimate subjects to the stairs, and in no trials were gross changes in normal gait observed.

**Hip adduction angle.** Hip adduction angle, defined as the relative angle between the pelvis and thigh, was higher in PFP subjects compared to pain-free subjects, but this observation was not significant (p=.10). As with knee valgus angle, high inter-subject variability among the pain-free subjects (M 1.5, SD ± 7.9 degrees; M 1.2, SD ± 3.9 degrees, respectively for peak and average angle) accounted for the lack of significance.

Additionally, PFP subjects had greater pelvis obliquity angle ("drop") toward the stance limb, effectively reducing their hip adduction angle, while pain-free subjects maintained a relatively neutral pelvis, increasing their hip adduction angle. Had hip adduction been measured using the thigh compared to a laboratory coordinate system (an unconventional measure) far greater differences would have been seen between groups.

The trend toward greater hip adduction angle lends some support to the "top-down" model for knee malalignment where increased knee valgus angle originates from hip alignment in the frontal plane (Powers, 2003).
**Essential Kinematic Discussion**

*Pelvis obliquity ("drop") angle.* Pelvis obliquity was defined as motion in the frontal plane, with reference to the laboratory coordinate system, where the pelvis would either tilt toward the stance limb (ipsilateral obliquity) or toward the opposite limb (contralateral obliquity). A significant difference was seen between PFP subjects and pain-free subjects for both peak and average obliquity angles, with PFP subjects having, on average, an ipsilateral peak drop of several degrees while pain-free subjects maintained a more neutral or contralateral drop (Figure 19).

The finding of increased pelvic obliquity among PFP subjects is not new. Mascari, et al (2003) described in a two-case series increased pelvis obliquity among PFP subjects when stepping down from an 8-inch step, however, they described a *contralateral* pelvis drop (Figure 32) instead of an ipsilateral drop. In this study, PFP subjects had a consistent ipsilateral drop, while pain-free subjects maintained a more level pelvis.

What remains unclear is the cause or causes of the observed pelvic girdle drop, and the differences in direction of the drop between studies. One theory suggests that pelvis obliquity may be attributed to weak hip abductors. Perry (1996) describes both ipsilateral and contralateral drops as accommodations to reduced hip abductor strength. The ipsilateral drop is referred to as the "compensated" or gluteus medius lurch gait, while the contralateral drop is referred to as an
"uncompensated" or Trendelenburg gait. In the former, the inability of the hip abductors to overcome the hip adduction moment during stance causes a "drop" of the pelvis to the same side, while the trunk moves ipsilaterally, effectively moving the center of mass over the stance limb (Figure 33.) In the latter, the pelvis drops opposite the stance limb, but still assumes a more ipsilateral position, again reducing the hip adduction moment (Figure 34). In both cases, strategies involving alteration of pelvis
and/or trunk alignment are thought to occur as compensations for weak hip abductors.

![Shift in center of mass.][1]

**Figure 34.** Ipsilateral pelvis obliquity with resulting thigh adduction and trunk lean to reduce hip adduction moment.
Figure 35. Contralateral pelvis drop with resulting thigh adduction and pelvis lean to reduce hip abduction moment.

However, in this study, significant differences in trunk side-bending were not observed, so reduction of hip moments did not necessarily occur. This will be discussed further in Essential Kinetic Discussion, below.

Alternatively, the tendency for the pelvis to drop to either side may result from weakened lumbopelvic muscles, such as the quadratus lumborum and abdominal musculature. As noted above, Mascal et al., (2003) demonstrated changes in pelvic girdle alignment during a step-down task in a series of two subjects by strengthening both the hip and the lumbopelvic muscles. Citing the work of Nedler, et al. (2002), who
demonstrated an association between lumbopelvic stability, low back pain, and hip strength, they (Mascal, et al), reasoned that: "Such stability is thought to promote greater torque production by these (hip abductor) muscles during exercise and minimize frontal plane motion..."

**Trunk flexion angle.** It was hypothesized at the outset that subjects with patellofemoral pain would exhibit increased trunk flexion in an attempt to move the center of mass of the superincumbent weight anterior to the knee to lessen the knee flexion moment. This was not observed, with PFP and control subjects showing no differences in peak and average trunk flexor motions. Part of the explanation for this may be the relatively limited sensitivity of trunk modeling in the sagittal plane accomplished with this study. By using four (and eventually five) cameras within a small laboratory space required positioning the cameras predominately in the frontal plane. As such, no posterior trunk markers were used, making three-dimensional measures in the sagittal plane difficult. This configuration did, however, produce reliable data for analysis of trunk motion in the frontal plane (side-bending) but somewhat questionable modeling of flexion and extension motions. In any case, the hypothesis of increased trunk flexion to reduce knee moments in PFP subjects was not supported.
Trunk side-bending angle. As with trunk flexion motion, it was hypothesized that PFP subjects would side-bend to the ipsilateral side to lessen the joint moments of the stance limb. This was not observed, however, with PFP and pain-free subjects having essentially the same side-bending tendencies.

This finding might partially explain the failure of PFP subjects to reduce their hip abduction moment during stance phase, as predicted by Perry (1996) in the strategies described above. (As discussed below, PFP subjects actually had higher hip abduction moments than control subjects, suggesting adjustments in the pelvis and trunk were insufficient to overcome hip abductor weakness.) Had subjects side-bent their trunks over the ipsilateral limb reduced hip moments would have been seen in this group. Failure to employ this strategy might explain the higher hip moments, and possibly the increased pelvic obliquity angles observed.

(A more complete discussion of the interactions of the kinematic and kinetic findings is developed below in the section subtitled “Essential Discussion- Correlation of Kinematic and Kinetic Findings,” page 102.)

General Kinetic Discussion

Knee flexion moment. As hypothesized, PFP subjects had significantly less peak and average knee flexion moments than did pain-free subjects. (Figure 23). This is consistent with the work of Salsich, et al. (2001) where individuals with patellofemoral pain descended stairs with
reduced knee moments (referred to in their work as internal "extensor" moments) compared to pain-free subjects (Figure 36). In a subsequent study (Salsich, et al., 2002), reported increased subjects' joint reaction force after application of patellofemoral taping as a pain-relieving modality. One difference in this study is that the initial peak of the bi-phasic moment came later among PFP subjects than pain-free subjects (Figure 22), which was not seen in the work of Salsich, et al. (2001). Otherwise, the general shapes and differences between groups are remarkably similar in these two studies.

Figure 36. Comparison of PFP and control subjects, from Salsich, et al. (Reprinted from Salsich, G., et al., Clinical Biomechanics 16(11), p. 910, 2001, with permission.)
The finding of reduced knee moments in subjects with patellofemoral pain is consistent with the "quadriceps avoidance gait" seen in individuals with knee pain during extensor mechanism loading (Perry & Fontaine, 1995). The altered relationship of the patella and femur in the PFP population reduces the contact area between the joints, resulting in greater joint stress (joint reaction force/joint surface area) and pain. (Brechter & Powers, 2002). Since these individuals are in pain, they will often find strategies to reduce joint moments ("quadriceps avoidance gait"). Thus reducing painful tasks (Powers & Landel, 1996). These strategies might include reducing knee flexion angle (as noted in this study and discussed above), reducing the velocity of stepping (not measured in this study), or altering their superincumbent center of mass by movement of the trunk (measured in this study but not found to be significant.)

Essential Kinetic Discussion

Knee valgus/varus moment. An important finding of this study is the significant differences seen between groups for peak and average knee valgus/varus moments, with PFP subjects having higher valgus values. In fact, pain-free subjects had slightly varus peak moments, while those with patellofemoral pain had consistent peak valgus moments (Figure 25). While joint angle and joint moment findings are not interchangeable measures, forces (moments) that act upon a joint must be balanced by an equal and opposite muscle force to maintain joint
equilibrium. In subjects in which an external knee valgus moment occurs, an internal (muscle) joint moment must be generated by the medial (pes anserine) muscles to balance the joint. Any weakness in the musculature required to balance the external force will result in valgus joint motion.

The finding of greater knee valgus moments among PFP subjects partly supports the observation that patients with patellofemoral pain will show biomechanical differences at the knee compared to those who are pain-free, of which greater knee valgus angles may play a contributing role (Powers, 2003).

Support for this premise is found in the literature. Etahi, et al. (2000) correlated patellofemoral osteoarthritis with increased valgus knee alignment (p=0.0066). In a follow-up study, Cahue, et al. (2004) found a high correlation between lateral patellofemoral changes and the 18-month progression of valgus knee angles. Finally, Fukui, et al. (2003) described a stress radiography technique in which subjects with suspected patellofemoral instability and normal subjects were radiographed with a mechanical valgus force applied to the knee and then compared to traditional Merchant views. Significant differences in patellofemoral congruence angles were seen between the groups for both techniques, but with a significant correlation (p=0.001) found between the stress x-rays and functional scores (measures of pain and functional ability) for the instability group.
Hip adduction moments. Both peak and average hip adduction moments were greater in PFP subjects than pain-free subjects (Figure 28). While it was suggested at the outset that PFP subjects would flex and/or side-bend the trunk in an attempt to accommodate knee pain or hip weakness, no significant differences were seen for trunk motions and moments. This may partly explain the higher hip adduction moments found among PFP subjects rather than the reduced hip moments expected.

Medial, frontal plane forces need to be balanced by an equal and opposite internal hip abductor moment (i.e. muscle moment) if a stable femur is to be maintained. By contracting eccentrically the hip abductors will control or limit the amount of hip adduction, while a concentric contraction (if the muscles are sufficiently strong) will cause the hip to actually abduct. (Further discussion of kinetic and kinematic interaction is developed below in Essential Discussion: Correlation of Kinematic and Kinetic Findings.)

Increased hip adduction motion can lead to increased knee valgus moment and angle. This is due to the fact that adduction of the femur will require a compensatory knee valgus (tibial abduction) motion to occur at the knee to allow the ankle and foot to assume a neutral position during stance phase of gait. While joint moments and joint angle findings are not interchangeable, any tendency for the knee to move toward valgus due to an adduction joint moment from above must be counterbalanced by an
internal muscle force, i.e. the pes anserine, otherwise increased knee
vagus motion will result.

Support for this premise can be found in the literature review, as
described above. Shim, et al. (1997) and Hefti (2000) correlated coxa vara
deformity (hip adduction angle) with genu valgum (increased knee valgus
angle). While their study dealt with fixed, bony, congenital deformities, the
compensatory mechanisms are the same for dynamic joint motions.

Essential Discussion- Correlation of Kinematic and Kinetic Findings

A significant contribution of this study is the correlation found
between pelvis obliquity angle and knee moments in the frontal plane for all
subjects, with patellofemoral patients having the greatest pelvis obliquity
angles and greatest knee valgus moments. As noted in the Results section
above, a correlation statistic was applied to the peak pelvis obliquity angle
and peak knee valgus moment, with a significant positive correlation found
(increasing pelvic obliquity toward the stance limb correlated with increased
knee valgus moment of the stance limb). This finding is important in that it
lends some support to the "top-down" theory of altered knee biomechanics
in PFP having a contribution from proximal biomechanical influences.

Shown below are two skeletal animation frames, the first (Figure 37)
depicting a control subject with a level pelvis and mild knee valgus
alignment and varus knee moment. The second (Figure 38) depicts a PFP
subject with ipsilateral pelvis obliquity, and increased knee valgus angle and valgus knee moment.

Figure 37. Control subject showing level pelvis, mild knee valgus angle, varus knee moment.
Figure 38. PFP subject showing ipsilateral pelvis obliquity, increased knee valgus angle and valgus moment.
One possible explanation for this finding is that weak hip abductors require a compensation that results in a pelvis drop toward the ipsilateral side during single limb support (compensated Trendelenburg gait, as discussed above.) Alternatively, weak pelvis stabilizing ("core") musculature may allow the pelvis to drop rather than maintaining a stable platform from which the hip abductors can work. In either case, the net result is increased thigh adduction and knee valgus moments.

While this correlation between pelvis and knee corroborates clinical observations (McConnell, 1998) and a 2-case report (Mascal et al., 2003) it does not take into consideration the "intermediate" joint of the hip acting between the pelvis and knee. As noted above, both hip adduction moment and knee valgus moment were, in themselves, significantly different between groups. However, a comparison between hip abduction moment and knee valgus moment curves for each group (Figures 26 and 29, respectively) fails to suggest a direct relationship. While the control subjects' hip abduction and knee varus moments produce mirror images of each other, the PFP subjects' knee popliteus (which are quite small, with a consistent valgus moment toward terminal stance,) bear little relationship to its hip adduction moment. Likewise, the differences between subjects for hip abduction moment (Figure 29), while statistically significant ($p=.001$), appear subtle in the graphic illustration, while the graphic depiction of differences in knee valgus moments (Figure 26), also significant at $p=.006$, are strikingly different between groups. Therefore,
further discussion on the interdependence of joint kinematics and kinetics is warranted.

Interdependence of Joint Kinematics and Kinetics

Initial review of the kinematic, top-down graphs (Figure 31) suggests an argument can be made that greater ipsilateral pelvis obliquity leads to greater hip adduction, which leads to greater knee valgus. However, these graphs only depict the mean of each group without considering the inter-subject variance. As noted previously, only pelvis obliquity angle had statistically significant differences between groups. Therefore interpretation of these mean graphs should be approached with caution.

Additionally, as was discussed regarding correlation of hip and knee moments, a logical flow of joint moments leading from one joint to the next was not seen (Figures 26 and 29.). Therefore, possible explanations for these findings are required.

One explanation for these findings is that joint moments occurring above and below a selected joint are just two of many factors that will influence the "middle" joint's mechanics (Winter, 1990.) As such, Winter states; "...moment of force curves should not be 'ooked at in isolation, but rather as part of a total integrated synergy in a given movement task." (1990, p. 92.)
Therefore, integrated biomechanical scenarios need to be offered to explain the findings of this study. In one scenario, based on the work of Perry (1990) and described above, ipsilateral pelvis drop and trunk lean would occur as a compensation for weak hip abductors, reducing hip adduction moment, as diagrammed in Figure 39.

Figure 39. Theorized biomechanical scenario to compensate for weak hip abductors.
As noted above, however, increased ipsilateral trunk lean was not seen, and hip adduction moments were actually higher. Therefore the biomechanical compensation proposed by Perry “failed” in the PFP group. As such, an increase in hip adduction moment occurred, resulting in increased femoral adduction, sometimes referred to as “thigh collapse” (Powers, 2003). This medial displacement of the distal femur likewise moves the knee joint center to a more medial position, increasing knee valgus moment and angle. This scenario is diagrammed in Figure 40.

Figure 40. Speculated scenario, where failure to compensate for hip abductor weakness causes increased hip adduction moment, increased knee valgus moment.
This scenario more fully explains the seemingly disparate findings of this study, although other factors also need to be considered. One consideration is that ipsilateral pelvis drop may have reduced an exceptionally large hip adduction moment among PFP subjects, but not enough to reduce it to that of control subjects. The remaining hip adduction difference would then be sufficient to cause an increased knee valgus among PFP subjects.

Another factor, not included in the above scenarios, is the pelvis width of individual subjects. While not specifically measured as a subject descriptor or dependent variable, subjects in the PFP group may have had larger pelvis widths (distance from ASIS to ASIS) than did control subjects. This variation, if present, would have influenced the lever arm of the hip abductors by moving the muscle attachment further from the hip joint center, resulting in greater hip adduction moments for this population (Horton & Hall, 1989). Since this dimension was not measured in subjects, the influence of pelvis width is, at this point, speculative.

Alternatively, patellofemoral pain may be “driving” the biomechanics. As noted above, this study, as well as that of Salsich and Brechtel (2003) found reduced knee flexion angles and moments among PFP patients as an apparent sagittal plane compensation to avoid pain. Possibly, PFP subjects would increase knee varus angle and moment in an attempt to unload the knee extensor mechanism by shifting forces to the medial knee stabilizers in the frontal plane, i.e., the pes anserine
muscle group. This would then suggest a frontal plane compensation to unload the painful extensor (patella) load. Figure 41 shows the theoretical basis for this premise, while Figure 42 shows anatomically how this "shift" in force attenuation might occur.

Figure 41. Alternative scenario: PFP drives biomechanical differences between groups.
Figure 42. Frontal plane compensation to unload painful extensor mechanism.
A final consideration of this “integrated synergy” is influence from the foot and ankle from below. While attempts to correlate foot and ankle mechanics to patellofemoral pain have not been successful (Messier, et al., 1991; Powers, et al., 2002) a theoretical framework exists (Tiberio, 1987) to suggest that altered foot mechanics may influence knee mechanics separate from patellofemoral pain reports. Since foot and ankle kinematics and kinetics were not measured in this study, their influence can only be speculated (Figure 43.)

![Figure 43. Bottom-up scenario, suggesting foot pronation influences knee biomechanics from below.](image-url)
As noted above, the "bottom-up" scenario has a compelling theoretical basis (Tiberio, 1987) but lacks biomechanical corroboration (Messer, et al., 1991, Powers, et al., 2002). Anecdotally, the subject depicted below (Figure 44) shows considerable knee valgus angle, as well as pelvis obliquity, yet examination of the foot reveals a neutral foot posture and vertical shank alignment, casting doubt on any "bottom-up" influence for this subject.

Figure 44. Neutral foot posture, vertical shank, despite increased knee valgus angle.
Summary

In summary, significant differences in the kinematics of the pelvis and kinetics of the hip and knee were found between women with and without patellofemoral pain when descending stairs, with a correlation found between frontal plane motion at the pelvis and frontal plane moments at the knee. While correlations of measures do not equate to cause and effect, the association between pelvis and knee lends some support for the "top-down" theory of knee mechanics having a proximal biomechanical influence.

Limitations of the Study

This study has several limitations. First, hip abduction strength was measured using traditional manual muscle testing, a common clinical tool, but one that lacks the reliability and validity required of formal research. For this reason, results of hip strength testing were used in the description of subjects, but not as a dependent variable for statistical analysis. As such, the argument that the biomechanical differences seen between groups is attributable to hip abductor weakness is speculative.

Another limitation of the study is the small sample size of groups (n=10 for both groups). While this sample size was sufficient to show significant differences in several of the independent variables in this study, as well as in other patellofemoral pain studies (Brecht & Powers, 2002), it was insufficient to produce significant differences across several more
variables, including hip adduction and knee valgus angles, two measures of primary importance to this study.

A third limitation, related to the first two above, is the broad admission criteria used in selecting patellofemoral pain subjects. Specifically, anyone with patellofemoral pain (who did not have any of the exclusion criteria, which were sufficiently strict) was admitted to the study. Since patellofemoral pain patients with proximal or "core" biomechanical differences may represent a subset of patients (Powers, 2003,) then a screening device (such as a reliable and valid hip abductor strength measure) could have been employed to pre-select subjects likely to have the suspected characteristics.

Finally, data collection for this study was done in a newly developed and still evolving biomechanics lab. Therefore, some segment modeling, including trunk flexion and extension (as discussed above) and shank and femoral rotation (not reported) was less than optimal, particularly when using a four-camera system. This limitation is not unique to this lab, however. As noted in the Literature Review, Yu and Kinebacher (1997) reported as much as 41% error in femoral rotation when using retro-reflective markers compared to bone pins. They attributed this error to skin and muscle movement of the markers compared to the "gold-standard" bone pins.
CHAPTER VI
CONCLUSION

Clinical Implications

Practitioners and researchers involved in the conservative management of patellofemoral pain are witnessing a "paradigm shift" in the theoretical basis of the etiology of this malady, with attention shifting from control of the patella atop the femur, to control of the femur beneath the tethed patella. While gaining in acceptance, this theoretical framework has not undergone rigorous basic science or clinical validation studies (Simoneau, 2003).

The purpose of this study was to investigate whether differences exist in the kinematics of the knee, hip, pelvis and trunk, and in the kinetics of the knee and hip between subjects with and without patellofemoral pain. Information from this work might help in the understanding of how the proximal joints or "core" influence the pathomechanics of the knee.

An important new finding in this study was the differences seen in knee valgus moments and pelvis obliquity angles between groups. While other researchers have suggested differences in the "core stability" of individuals with PFP (McConnell, 1986, Powers & Silvers, 2002) leading to malalignment of the knee, no known studies have examined the differences in knee moments and pelvis angle by collecting data on multiple subjects.
The finding that PFP subjects in this study consistently "dropped" their pelvis toward the stance limb during descent of stairs, while pain-free subjects maintained a level or elevated pelvis, lends some credibility to the "top-down" theory of the origin of knee malalignment and subsequent pain having a proximal biomechanical origin (Powers, 2003).

The most likely explanation for the observed drop of the pelvis is hip abductor weakness causing either ipsilateral pelvis obliquity compensation, as seen in this study, or contralateral pelvis obliquity, as seen in previous work (Mascal, et. al., 2003). Therefore, addressing hip abductor strength, as suggested by Ireland, et al (2003) may be sufficient to address motion at the pelvic "base."

Alternatively, motion at the pelvis may also be caused by weak pelvis stabilizing or "core" muscles, as suggested by Nadler, et al. (2002) and Leetun, et al. (2004) and referenced by Mascal, et al. (2003) as a theoretical basis for their intervention.

This interpretation suggests that rehabilitation programs aimed at correcting biomechanical malalignment should include hip abduction strengthening exercises, as well as exercises to maintain pelvis or "core" stability to provide a firm base from which the hip abductor muscles can work. This would include strengthening of the muscles joining the pelvis and trunk, including the back extensors, abdominals, and quadratus lumborum (Leetun, et al., 2004).
Suggestions for Further Research

1. Given the relatively small sample size used in this investigation, repeating the study with larger groups seems warranted. In particular, post-hoc power analysis of the trend toward significance for knee valgus angle and hip adductor angle suggested that the addition of 5 to 10 more patellofemoral pain subjects (n = 20) would have demonstrated significance for these important measures.

2. Since hip abduction strength appeared to play a significant role in the theoretical interpretation of study findings, future work should include a valid and reliable measure of hip abductor strength as an independent variable for subjects. A correlation between hip abductor strength and pelvis and knee motions would help solidify the "top-down" theoretical framework.

3. Since patients with patellofemoral pain and hip abductor weakness appear to represent a subset of the PFP population, a clinically-based screening tool needs to be found to identify these patients. A study demonstrating the validity and reliability of such a tool might lead to a classification system of PFP patients, as has been called for by several researchers and clinicians (Powers, 2003; Simoneau, 2003; Bizzini, et al., 2003).

4. Most importantly, the finding of altered pelvic girdle motion among PFP patients in this study, which appears to be amenable to
corrective exercises (Mascal, et al. 2003,) suggests that a pre- and post-exercise (hip and/or pelvic girdle strengthening program) study of pelvis and lower extremity kinematics and kinetics with multiple PFP subjects is needed.

5. Finally, testing the alternative hypothesis that PFP "drives" the biomechanical differences between groups could be investigated by comparing kinematic and kinetic changes following application of a pain relieving modality, such as patellofemoral taping. This intervention has proven remarkably successful in relieving symptoms, as well as changing biomechanical measures (Powers and Landel, 1997.) Significant changes in biomechanics following tape application would suggest a "pain driven" biomechanical accommodation, while failure to change would suggest biomechanical differences from a separate cause.
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Appendix A
Solicitation Flyer
DO YOU HAVE KNEE PAIN WHEN DESCENDING STAIRS?

Professor Jim Phillips in the Department of Physical Therapy and Sports Science in the School of Graduate Medical Education is conducting a study on hip strength and the way people descend stairs.

Information from this study may help in our understanding of why some people have knee pain when descending stairs and may lead to exercises that can correct the problem.

Participation requires one visit to the motion analysis lab in Duffy Hall for approximately 2 hours. You will undergo simple hip muscle strength testing, then descend several steps while wearing reflective markers that specialized cameras will “see” to record your steps.

Your participation is completely voluntary, and your identity will be kept strictly confidential, with all personal data kept in coded form.

Interested? Contact Professor Phillips at 201 370 7195, or at phillih0@shu.edu

This project has been reviewed and approved by the Seton Hall University Institutional Review Board for Human Subjects Research. The IRB believes that the research procedures adequately safeguard the subject’s privacy, welfare, civil liberties, and rights. The Chairperson of the IRB may be reached at (973) 275-2974.
Appendix B

Letter of Solicitation
Dear Doctor:

I am a doctoral student and Assistant Professor in the School of Graduate Medical Education at Seton Hall University. I am conducting a study on the influence of hip strength on patellofemoral pain and knee kinematics (motion) and kinetics (force) during descent of stairs. Information from this study may help in our understanding of why people have knee pain when descending stairs and may lead to exercises that can correct the problem.

Qualification for the study includes the following: anterior knee pain during descent of stairs, and any one or more of the following: a) anterior knee pain following prolonged sitting, b) anterior knee pain during ascent of stairs, c) anterior knee pain during kneeling, d) anterior knee pain during running, d) anterior knee pain during jumping, and e) anterior knee pain during isometric quadriceps contraction. Exclusion criteria includes each of the following: 1) any prior surgeries to the symptomatic knee, 2) any neurologic condition affecting the lower extremities that would influence gait, 3) any ligamentous or meniscal instability of either knee that would influence gait, and 4) history of traumatic patellar dislocation.

Participation requires one visit to the motion analysis lab in Duffy Hall at Seton Hall University for approximately 2 hours. Subjects will undergo simple hip muscle strength testing, then descend several steps while wearing reflective markers that a specialized camera will "see" to record their steps.

Your patient’s participation is completely voluntary, and their identity will be kept strictly confidential, with all personal data kept in coded form in a locked cabinet.

If any of your patients are interested in participating, please have them contact me at 973 275 2250, or by e-mail at philibo@shu.edu.

Thank you for your interest and assistance in this project.

Sincerely,

H. James Phillips, PT, OCS, ATC

This project has been reviewed and approved by the Seton Hall University Institutional Review Board for Human Subjects Research. The IRB believes that the research procedures adequately safeguard the subject’s privacy; welfare, civil liberties, and rights. The Chairperson of the IRB may be reached at (973) 275-2974.
Appendix C

Informed Consent Form
Informed Consent Form

"Effect of hip strength on patellofemoral pain, kinematics, and kinetics during descent of stairs."

Investigator:
This study is being conducted by H. James Phillips, doctoral candidate and Assistant Professor in the School of Graduate Medical Education of Seton Hall University.

Purpose and Duration of Study:
The purpose of the study is to see whether people who experience knee cap pain when descending stairs have hip weakness compared to people without knee pain, and to see whether they descend stairs differently to help lessen the pain. The study requires two visits to Seton Hall University, the first lasting about 15 minutes to see if you qualify, and a second visit of no more than 2 hours for testing.

Procedures:
During the first visit you will answer a simple questionnaire about how your knee feels for certain daily activities. If you meet certain criteria, you will then have your knee evaluated by the investigator who is a licensed physical therapist and certified athletic trainer to make sure there is no ligament or other knee injury that would prevent you from participating. None of these tests are strenuous or uncomfortable. This session takes about 10-15 minutes. If you qualify for the study you will be asked to return another day for the actual experiment which takes no more than 2 hours.

The experiment begins with you walking down stairs, then marking a scale to show if and how much knee cap pain you are experiencing. The next part of testing involves simple hip muscle strength testing involving lifting your leg against gravity, or with some manual pressure from the investigator. This testing is not strenuous or painful. For the rest of the experiment you will be asked to wear form fitting, bicycle-type shorts, and a sports bra or swim top. This is so reflective markers can be placed over various parts of your trunk, pelvis and legs so they will move as you move. You will then be asked to walk down three regular steps while a force plate records how much weight you are placing on the steps, and motion cameras record the movement of the reflective markers. (The cameras do not take regular pictures that show your face or body). The total number of times you will walk down the steps will be kept to a minimum, totaling no more than 3 flights of stairs for the entire session.
Voluntary Nature of Project:
Your participation in the study is completely voluntary. You may stop participation at any time, even in the middle of testing, without any penalty or consequence.

Anonymity:
All recorded information from the study will be kept in coded form so no one but the investigator will know how you performed. All information will be kept in a locked cabinet in the investigator’s office. Only the investigator will have access to your data.

Risks:
There are no major risks in participating in this study. The only knee discomfort you will experience is the same as you would when walking down any flight of stairs. An elevator is available to take you from the motion lab on the second floor to ground level if you would like to avoid stairs after testing. There are activities that should produce any anxiety or stress before, during, or after testing.

Benefits:
Your participation in this study may assist in developing a better understanding of why people have knee-cap pain, and lead to ways of treating it. There is no financial compensation for participation in this study.

Alternate Therapies:
This experiment does not involve treatment of your knee condition. However, the primary investigator will provide advice, as permitted within his scope of practice as a licensed physical therapist, about things you can do for your knee.

Questions:
If you have any questions at any time, please contact the investigator, H. James Phillips in the Department of Physical Therapy and Sports Science, Allison Hall, Room 34; at 975 275 2250; or at phillih@eshu.edu.

A copy of this consent form will be provided to you.

This project has been reviewed and approved by the Seton Hall University Institutional Review Board for Human Subjects Research. The IRB believes that the research procedures adequately safeguard the subject’s privacy, welfare, civil liberties, and rights. The Chairperson of the IRB may be reached at (973) 275-2974.

I have read the material above, and any questions I asked have been answered to my satisfaction. I agree to participate in this activity, realizing that I may withdraw without prejudice at any time.

Subject __________________________ Date __________________________
Appendix D

Inclusion/Exclusion Questionnaire
Inclusion/Exclusion Questionnaire

Subject Number: __________

Do you have patella (kneecap) pain when descending stairs? Yes ___ No ___

Do you have patella pain with any of the following?

- When ascending stairs: Yes ___ No ___
- When kneeling: Yes ___ No ___
- When running: Yes ___ No ___
- When jumping: Yes ___ No ___
- After sitting for a long time: Yes ___ No ___
- When you tighten your thigh muscle: Yes ___ No ___
  (The investigator will help with this one).

Have you had any surgeries to your painful knee? Yes ___ No ___

Have you ever dislocated your painful kneecap? Yes ___ No ___

Are you aware of any neurologic conditions that effect your walking? Yes ___ No ___

Have you been told that you have ligament or knee meniscus (cartilage) injury of your knee? Yes ___ No ___

Qualifies for knee exam: _________ Excluded from knee exam: __________
Appendix E

Knee Examination Form
Knee Examination

Subject Number: __________

Ligamentous Testing

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<th>Negative</th>
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<td></td>
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<tr>
<td>Lachman’s Test</td>
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<td></td>
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<tr>
<td>Posterior Drawer</td>
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<tr>
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<td></td>
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<tr>
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<tr>
<td>Valgus Stress</td>
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Meniscal Testing

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<td>Bounce Home Test</td>
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Neuro Screening

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<th>Hyper</th>
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<td>Quadriceps Deep Tendon Reflex</td>
<td>______</td>
<td>______</td>
<td>______</td>
</tr>
</tbody>
</table>

Qualifies for study: __________ Does not qualify for study: __________
Appendix F

Knee Discomfort Form
KNEE DISCOMFORT WHEN DESCENDING STAIRS

Subject Number: __________
Study Entry Testing: _____ Pre-Experiment Testing: _____

Instructional Script:

"In a moment you will be asked to descend a regular flight of stairs. When finished, please mark the scale below with a pen indicating how much discomfort you experienced. The far left of the scale represents NO KNEE PAIN. The far right of the scale represents WORST IMAGINABLE KNEE PAIN. Mark the scale where your pain would fall."

________________________________________

"No knee pain"                               "Worst imaginable knee pain"

Knee Pain in cm: __________
Appendix G

Hip Strength Assessment Form
HIP ABDUCTION STRENGTH

Subject Number: _____________

<table>
<thead>
<tr>
<th>Grade</th>
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<th>Description</th>
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</thead>
<tbody>
<tr>
<td>Normal (5/5)</td>
<td></td>
<td>Completes full ROM against gravity and resists moderate manual pressure.</td>
</tr>
<tr>
<td>Good (4/5)</td>
<td></td>
<td>Completes full ROM against gravity and resists mild manual pressure.</td>
</tr>
<tr>
<td>Fair (3/5)</td>
<td></td>
<td>Completes full ROM against gravity, but cannot resist minimal manual pressure.</td>
</tr>
<tr>
<td>Fair minus  (3-5)</td>
<td></td>
<td>Completes partial ROM against gravity.</td>
</tr>
<tr>
<td>Poor (2/5)</td>
<td></td>
<td>Can only abduct leg with gravity eliminated.</td>
</tr>
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</table>
Appendix H

CalTester Report
Report Format

<table>
<thead>
<tr>
<th>Min (at x cm N)</th>
<th>Mean ± 1 Std. Dev.</th>
<th>Max (at x cm N)</th>
</tr>
</thead>
</table>

The mean, standard deviation, and range (minimum and maximum) of the report variables are provided on the graph and with numerical values. The applied force of the time of the minimum and maximum value is indicated alongside the value.

Summary of Differences

Δ Force Orientation (°) = 1.6 ± 0.7

Δ Force Orientation is the angle between the orientation of the axial component of the applied force and the orientation of the load axis of the device, as determined from the target data.

Δ CoP x (mm) = 8.2 ± 6.3

Δ CoP y (mm) = 9.2 ± 6.9

Δ CoP z (mm) = -23.6 ± 6.3

Δ CoP x, y, z are ab components of the displacement vector between the Center of Pressure location measured by the force platform and the endpoint of the calibration testing device (adjusted for a vertical height of 0.0 mm above the force platform.)

Report generated by CalTest (© C-Action Inc.)
Appendix I

CalTester Device
Cal Tester Device